Dynamic Control Strategies Used by ACL-Deficient Copers and Non-Copers during Stair Climbing

By

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final versions, as accepted by my examiners.

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Abstract

Background: Osteoarthritis (OA), the most common form of arthritis, is a disorder that affects bi-articular synovial joints and is characterized as the focal loss of articular cartilage with changes to the subchondral and marginal bone, synovium and para-articular structures. Biomechanical factors such as obesity, joint loading, joint injury, joint deformity, level of sports participation, occupational factors, and muscle weakness specifically affect joint loading which influences local OA development. One way to assess the effect of altered loading patterns on a joint is to examine changes following joint injury. An appropriate model for this purpose is the anterior cruciate ligament (ACL) rupture in the knee joint. Recent evidence has shown that some ACL-D subjects are able to perform activities at levels similar to those of normal subjects. Termed 'Copers', these individuals demonstrate greater knee stability than other ACL-D subjects who are considered 'Non-Copers'. Therefore, the purpose of this study was to evaluate the dynamic control strategies used by ACL-D Copers and Non-Copers during a stair climbing task, and compare them to a healthy population as a means to identify possible susceptabilities to OA.

Methods: A total of 20 participants were collected including 10 (5 male and 5 female) Healthy controls, 5 (2 male and 3 female) ACL-D Copers, and 5 (1 male and 4 females) ACL-D Non-Copers. The Healthy group had an average mass of 75.9 kg ±16.1 kg and an average height of 174.3 cm ±12.1cm, while the Copers group had an average mass of 71.7 kg ±6.2 kg and an average height of 170.8 cm ±5.6 cm, and the Non-Copers group had an average mass of 71.3 kg ±13.1 kg and an average height of 168.7 cm ±4.8 cm. ACL-D subjects were classified as either a Coper and Non-Coper based on a screening examination prior to the evaluation.The stair climbing task involved each subject completing 20 stair-climbing trials (10 ascent and 10 descent). Kinematic and kinetic outcomes were collected on the lower limbs, as well as,electromyography (EMG) of the tibilais anterior, medial and lateral gastrocnemius, vastus lateralis, vastus medialis, semitendinosus, and biceps femoris muscle group were collected bi-laterally. A one-way analysis of variance (ANOVA) was used to test each outcome's statistical significance ($\alpha = 0.05$).

Results and Discussion: During ascent the Non-Copers and Copers group completed the task with greater knee flexion than the healthy group (although not significant), while during descent the Non-Copers maintained approximately 5 degrees greater flexion than the other groups. A significant difference was observed with the moment about the knee (flex-ext) between the Healthy group and the Non-Copers group during the late stages of stair ascent with the healthy group maintaining a flexor moment and the Non-Copers group using an extensor moment. This extensor moment correlated with the increase of quadriceps activity among the Non-Copers which may lead to instances of knee joint instability. The Non-Copers group had increased activity of the hamstring muscle group without any attenuation in the quadriceps muscles. This will most likely increase the forces acting on the knee joint. Since, the knee is at greater flexion with an increased amount of forces, the Non-Copers may be at a greater risk articular cartilage breakdown. Throughout the majority of descent we saw an elevated amount of hamstring activity in the Non-Copers group at greater knee flexion angles as a potential means of controlling displacement of the tibia. This increase in hamstring muscle activity by the Non-Copers, without any significant decrease in quadriceps activity, places greater forces on the knee at different knee angle, when compared to the healthy group and may promote the onset of osteoarthritis. The Non-Copers also had increased moments about the knee joint for the majority of stance in both the flexion-extension plane and the abduction-adduction plane, when compared to the healthy group.

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1. Introduction

Osteoarthritis (OA), the most common form of arthritis, is a disorder that affects bi-articular synovial joints and is characterized as the focal loss of articular cartilage with changes to the subchondral and marginal bone, synovium and para-articular structures (Rottensten, 1997). The Arthritis Society (2005) estimates approximately 1 in 10 Canadians are afflicted with OA; that is 4 million Canadians. In the United States of America (USA), OA of the knee joint, one of the most common joints affected, occurs in approximately 6% of the population over the age of 30 and upwards of 30% of the population over the age of 65 (Felson et al., 1990). Due to the high prevalence of knee OA there is a need for research in the area of OA prevention.

The development of osteoarthritis is attributed to several risk factors including both systemic and biomechanical factors. Systemic factors that influence the likelihood of developing osteoarthritis include age, sex, ethnic characteristics, bone density, estrogen replacement, nutritional factors, and genetics (Felson, 2000). These factors affect the general susceptibility of developing osteoarthritis. Biomechanical factors such as obesity, joint loading, joint injury, joint deformity, level of sports participation, occupational factors, and muscle weakness specifically affect joint loading which influences local OA development (Felson, 2000). For instance, for each ½ N increase in weight, the overall force across the knee in single-leg stance increases 9-13 N (Felson, 2000).

The structure and function of articular cartilage is primarily determined during childhood as a result of loading conditions (Arokoski et al., 2000). During adulthood, normal physiological loading helps to maintain the integrity of articular cartilage tissue (Arokoski et al., 2000). This loading is essential to prevent atrophic changes in articular cartilage tissue (Palmorski & Brandt, 1981). Early stages of OA are characterized by focal and progressive hyaline cartilage loss with concomitant changes in the bone underneath the cartilage, including marginal outgrowths, osteophytes, and increased thickness (Felson, 2000). Following joint injury, the articular cartilage becomes thicker, softer, and more permeable, thus affecting the material properties of the articular cartilage (Wu et al., 2000).

Normal joint loading is an important variable in the maintenance of articular cartilage health. An abnormal increased joint loading affects the development of OA; however before we can determine how joint loading can affect OA development we first need to further understand the mediators of joint loading. One way to assess the effect of altered loading patterns on a joint is to examine changes following joint injury. An appropriate model for this purpose is the anterior cruciate ligament (ACL) rupture in the knee joint. According to a review by Yu et al. (2002), it has been estimated that 1 in 3,000 people in the USA have an ACL rupture. Animal studies by Hasler et al. (1998) and Herzog et al. (1998) where the ACL was transected in cats showed material property changes within the articular cartilage and altered joint loading patterns resulting in the initiation of arthritic changes. Therefore, the ACL ruptured knee is a good model for studying how an alteration in joint loading may lead to OA.

The primary role of the ACL is to prevent anterior translation of the tibia in relation to the femur when an anterior shear force is applied to the knee. When the ACL is compromised, the anterior-posterior stability of the joint is reduced and the loading conditions of the joint are altered. These changes include alterations in load magnitude, direction and load distribution over the joint surface. These changes may lead to the degeneration of articular cartilage tissue. Thus, ACL injuries provide a means for investigating how individuals adapt to altered loading conditions and how these loading conditions may affect the integrity of the articular cartilage.

Studies have shown a large variability in the movement patterns of ACL-deficient (ACL-D) subjects (Berchuck et al., 1990, Roberts et al., 1999). Recent evidence has shown that some ACL-D subjects are able to perform activities at levels similar to those of normal subjects. Termed 'Copers', these individuals demonstrate greater knee stability than other ACL-D subjects who are considered 'non-Copers' (Rudolph et al., 1998).

In 2003, Alkjaer et al. evaluated the walking pattern of ACL Copers and non-Copers and found that Copers exhibited a greater peak knee flexion angle during stance than non-Copers, while the controls exhibited larger extensor moments for a given peak flexion angle when compared to Non-Copers. This study concluded that all 3 groups walked with a different pattern and that Copers tended to stabilize the knee joint by co-contracting the hamstrings and quadriceps muscle groups at key instances where knee joint stability needed to be optimized while the non-Copers did not. The non-Copers did however prevent anterior displacement of the tibia by reducing the knee extensor moment. Using different strategies to maintain dynamic stability of the knee during walking produces different shear and compressive load demands on the joint and therefore alters the risk of developing OA.

Static stability of the knee is mediated by the integrity of the passive structures such as the ligaments and joint capsule. A decrease in static stability is often referred to as joint laxity. Conversely, a large portion of dynamic stability is the result of the coordination of muscle activation patterns in order to stabilize the joint. Copers have been shown to display kinematics and joint moments at the knee that are similar to healthy individuals. Copers also display less overall co-activation across the entire cycle than non-Copers (Rudolph et al., 2001). Non-Copers exhibit disturbed motor patterns that not only stiffen the knee at unwanted times but also delay muscle activations that are crucial for maintaining dynamic stability of the knee (Rudolph et al., 2001).

Activities of daily living (ADL) create mechanical stresses on the joints, which could be a source for development of OA especially following joint injury where the stresses at the joint are increased. A common ADL is stair ascent and descent. To date, there has not been any research into the differences between ACL Copers and non-Copers during stair ascent and descent. Stair climbing imposes a more challenging environment for the knee joint when compared to normal walking and thus may provide additional insight into the differences between Copers, non-Copers and healthy individuals. Knowledge of dynamic control strategies and techniques for stabilizing the knee used by Copers and Non-Copers during stair climbing maybe vital to understanding how the joint is loaded differently and how these differences may affect the risk of OA development. This knowledge can lend insight into rehabilitation methods to approximate normal loading conditions of the knee joint following joint injury in order to decrease the risk of OA. Therefore, the purpose of this study is to identify the dynamic control strategies used by ACL-D Copers and non-Copers to stabilize the knee joint during a stair climbing task. The following hypotheses will be examined specifically:

1) ACL-D Copers and Non-Copers will display different moments about the injured knee during stair ascent and descent; and
2) Copers will flex the knee to a greater extent, thus allowing the hamstrings to have a greater mechanical advantage to resist anterior tibial translation, while the non-Copers will reduce the extensor moment at the knee in an attempt to reduce anterior tibial translation; and

3) The Copers group will co-contract to a greater extent at specific instances during the stair climbing task while the Non-Copers will co-contract at lower levels for a greater duration of the stair climbing task.

2. Literature Review

2.1 Initiation and Progression of OA

OA is triggered via a mechanical overload that causes microdamage in the subchondral bone provoking a biological response that potentiates the progression of articular cartilage damage (Thambyah, 2005). Joint vulnerability or excessive loading is thought to create an environment that may trigger events leading to the development of OA (Thambyah, 2005). Joint injury or excessive loads placed on the joint may directly stimulate or result in damage or injury involving the subchondral plate and calcified cartilage near the tidemark, the junction of the non-calcified and calcified portions of mature articular cartilage. Damage to the subchondral plate or calcified cartilage triggers the enchondral ossification at the tidemark region causing a progressive thickening of the subchondral bone and thinning of the cartilage (Thambyah, 2005). In a ACL-D knee, chondral injury may be the result of repeated increased tibio-femoral translation in addition to episodes of 'giving way', that may increase the shear stresses on tissues in the knee joint causing further injury resulting in the progression of OA (Jones et al. 2003).

2.2 Development of OA as a result of ACL transection

There has been considerable research involving the progression of OA as a result of ACL transection. Most papers claim that following ACL transection there are changes in the material properties of the cartilage. There have been reports of changes in the knee joint within days of ACL transection. Some of these changes include site specific increases in cartilage thickness that results in an overall decrease in cartilage stiffness in both compression and tension (Herzog et al., 2003). Decreased stiffness is related to decreased amounts of proteoglycans and will cause the collagen fibres to break and fray. There is conflicting evidence as to how the articular cartilage becomes thicker following ACL transection. Setton et al. (1994) believed that the increase in cartilage thickness was due to an influx of water content in the cartilage, however, Herzog et al. (1998) disagrees since his experimental evidence shows the permeability of the cartilage does not change following ACL transection. There is little doubt that transection of the ACL affects of the material properties of the cartilage. What is mostly unknown is how altered loading affects the material properties of the cartilage and its ultimate degeneration. It is known that transection of the ACL alters the compressive stresses acting on the articular cartilage (Hasler et al., 1998). Following ACL transection, forces measured from the patellar tendon and gastrocnemius actually decreases within days (Hasler et al., 1998). This evidence suggests that the forces acting on the articular cartilage may decrease as well. Following ACL transection the forces acting on the cartilage decrease (Herzog et al., 1998). Measured in an animal model, Herzog et al. (1998) found that the peak pressure acting on the articular cartilage decreases but the overall contact area increases. As stated earlier, normal physiological loading helps to maintain the integrity of articular cartilage tissue (Arokoski et al., 2000). It seems that transection of the ACL alters the normal physiological loading and therefore, the integrity of the cartilage. Transection of the ACL produces joint instability resulting in an increase in overall contact area, causing loading of articular cartilage surface locations that are typically unloaded in the normal knee. Therefore, specific locations of the articular cartilage surfaces are overloaded while other locations are underloaded following transection of the ACL (Wu et al., 2000). This overloading of specific regions within the joint may elicit the onset and progression of degenerative changes despite a general unloading of the joint measured by reduced peak contact pressures, reduced gastrocnemius forces, and reduced patellar tendon force (Wu et al., 2000, Hasler et al., 1998).

2.3 Knee Motion of ACL Deficient (ACL-D) subjects during gait

An area of investigation is the study of knee motion during activities of daily living (ADL). During ADL, ACL-D subjects must dissipate the loads that are normally carried by the ACL to other joint structures and are at a higher risk of developing osteoarthritis (Hoffman, 1993). This is due to the abnormal stresses placed on articular cartilage within the knee joint. The most common ADL is that of locomotion.

A significant amount of research has been given to the adaptation and compensatory mechanisms used by ACL-D subjects in order to walk. During locomotion, the force that is typically transmitted through the ACL depends on the shear forces that act on the knee (Shelburne et al., 2004). The ACL is typically loaded whenever the net shear force at the knee is in the forward direction. During walking, the largest amount of anterior shear force at the knee is during the early stance phase where the greatest amount of anterior shear, of the tibia with respect to the femur, is caused by the patellar tendon transmitting force from the quadriceps muscle group (Shelburne et al., 2004). An educated assumption would be that during walking, ACL-D subjects would try to minimize the amount of anterior shear force at the knee, by limiting the amount of quadriceps muscle force acting through the patellar tendon.

One of the most widely used terms to describe the gait of ACL-D subjects is 'quadriceps avoidance'. This was a term used by Berchuck et al. (1990) to describe their findings of 12 out of 16 ACL-D subjects while they performed activities such as level walking and jogging. This study found that during the heel strike phase of gait, ACL-D subjects had a significantly greater net extension moment. During the mid-stance phases of gait, normal subjects display a large net flexor moment at the knee. However, ACL-D subjects did not exhibit this large flexor moment; instead they produced an extension moment (Berchuck et al., 1990). From the results of this study it was evident that during stages of the gait cycle, ACL-D subjects reduced the flexion moment at the knee. Since, flexion moments at the knee are countered by activation of the quadriceps muscle it was evident that ACL-D subjects reduced or avoided instances where quadriceps activation was needed creating a 'quadriceps avoidance' gait pattern. One limitation of this study was that there was no recording of quadriceps muscle activation. Data from this study only came from using a force plate and a motion analysis system. Therefore, moments presented in this paper are only net moments and give little insight into how these moments were actually obtained.

Work done by Beard et al. (1996) set out to re-examine the strategies used by ACL-D individuals to maintain dynamic joint stability during walking by measuring the muscle activity through EMG. Results of this study showed that ACL-D subjects walked with greater knee flexion angles compared to their contra-lateral leg as well as when compared to control subjects. A key finding of this study was that during the stance phase of gait, where Berchuck et al. (1990) assumed that ACL-D subjects avoided quadriceps contraction, Beard et al. (1996) found that ACL-D subjects actually had a significant increase in hamstring muscle activity on the injured side when compared to controls without any decrease in quadriceps activity indicating a level of co-contraction. Since the work done by Berchuck et al. (1990) presented results only on the net moments of the knee, then it is a possibility that the net flexor moment at the knee is the result of increased hamstring activity rather than a quadriceps avoidance strategy. Similar results were reported by Kadaba et al. (1993), where they found that ACL-D subjects walked with a more flexed knee pattern with an increase in knee flexion moment which would counter the proposed ‘quadriceps avoidance’ compensatory mechanism proposed by Berchuck et al. (1990).

Roberts et al. (1999) examined the issue of ‘quadriceps avoidance’ during gait and found that none of the ACL-D subjects displayed a decrease in quadriceps muscle activity. This was a critical finding considering Berchuck et al. reported it in 75% of the patients tested.

This major difference may in part be due to time post injury. Birac et al. (1991) reported that patients who had a post injury time of around 6 years were more common to walk with a ‘quadriceps avoidance’ gait pattern. Patients from the Roberts et al. (1999) and the Kadaba et al. (1993) studies had similar post injury times for patients which may lead to the similar results. A common finding from these studies was that ACL-D subjects did not extend their knees as much as the control groups did during the stance phases of gait which coincides with previous studies reporting an increased duration of hamstring muscle activity during this phase. Having a more flexed position during the stance phase of gait allows the hamstring muscles to compensate for the deficient ACL (Roberts et al., 1999). The hamstring muscles are effective in reducing forces of the ACL at all flexion angles except for those at full extension (Pandy & Shelburne, 1997). Therefore, walking with a more flexed position and an increase in hamstring activity allows a more effective compensation for a deficient ACL.

There have been numerous studies that have evaluated the effect of ACL deficiency on sagittal plane kinematics but few examined the effect in the transverse and coronal planes. During knee flexion the tibia internally rotates with a ‘screwing home’ motion in normal knees. However, in the ACL-D knee the tibia internally rotates for only the first 30° of flexion and then externally rotates to approximately 60° and the internally rotates again

(Dennis et al., 2005). This reverse 'screwing home' technique observed in ACL-D subjects may increase the risk of knee instability (Dennis et al., 2005). Similarly, a study by Zhang et al. (2003) showed that ACL-D patients walked with a similar pattern of internal/external tibial rotation but an overall increase in external tibial rotation when compared to the control group. This difference was more prominent during the stance phase of gait. This study also showed few differences in the abduction and adduction of movements between ACL-D subjects and controls.

Collectively the results of these studies indicate that ACL-D subjects attempt to walk with a pattern of gait that typically unloads the ACL (increased knee flexion, increased external tibial rotation). There is still a fair amount of controversy regarding what is a typical ACL-D gait pattern. A possible explanation reported earlier is the amount of time post ACL injury did the testing take place. Another explanation to the conflicting theories of ACL compensation during gait is the uniformity of the participants within the ACL-D group. It has been previously stated that ACL-D subjects can be separated into 2 groups: Copers and non-Copers. The studies discussed thus far did not separate their participants into these 2 groups. The ratio of Copers and non-Copers in the subject pool could have vastly influenced the results. Only recently did Alkjaer et al. (2003) separate ACL-D subjects into groups of Copers and Non-Copers and evaluated their different walking patterns.

2.4 Gait differences between Copers and Non-Copers

Due to the fact that early research on gait patterns of ACL-D subjects provided conflicting evidence, there was a need to separate the ACL-D into groups depending on functional ability. Therefore, differences between ACL-D Copers and Non-Copers compared to normal individuals have been investigated (Alkjaer et al., 2003, Rudolph et al., 1998). Copers and Non-Copers were differentiated based on their scores on the Lysholm and Tegner scores with no functional tests.

In a study by Alkjaer et al. (2003), controls, Copers, Non-Copers did not show significant differences when considering stride characteristics (velocity, step length, cadence, swing time, stance time). However, when considering kinematic and kinetic variables, differences were seen among groups. There was a significant difference in knee flexion between the Copers and control, with the Copers group having larger peak knee flexion angles, 27.1° (sd=6.2) compared to 21.2° (sd=4.9). The Copers group also had larger knee flexion angles when compared to Non-Copers group, but this was not significant. Also, at a given knee flexion angle the knee extension moment was significantly greater in the controls than in the Non-Copers but not significantly greater than the Copers (Alkjaer et al., 2003).

Results from Alkjaer et al. (2003) indicate knee kinematic patterns are similar between the Non-Copers and controls and the knee extensor moments were similar between the Copers and controls. The Copers group was able to load the knee joint in a similar fashion compared to the controls by stabilizing the knee through increased hamstring activation. The Non-Copers group decreased the moments produced by the knee extensors, in an attempt to decrease anterior translation of the tibia. This produced differences in the joint loading profiles. However, similar knee joint kinematics, between the controls and Non-Copers, may mask the differences in knee moments produced by the knee extensors. Therefore, at a given knee joint angle the controls significantly loaded their knee more than the Non-Copers (Alkjaer et al., 2003). This study suggests that the differences seen in peak knee extensor moment from earlier studies are dependent on peak knee flexion and the group tested (Alkjaer et al., 2003).

Alkjaer et al. (2003) also measured the activation of the thigh musculature during gait and found that the Copers group tended to stabilize their knee joint through coactivation of the vastus lateralis (VL) muscle and the biceps femoris (BF). When looking at individual muscles the Copers group showed greater EMG amplitudes of the VL and BF over the other two groups while the Non-Copers group had smaller EMG amplitudes of the VL when compared to the controls but larger BF amplitudes (Alkjaer et al., 2003). Therefore, it seems the Non-Copers group displayed more of a ‘quadriceps avoidance’ gait than the Copers did.

This study illustrates that the uniformity of the ACL-D groups in the previous studies may have had an impact in the results. It is therefore necessary to distinguish between ACL Copers and ACL Non-Copers when performing research studies and evaluate their movement patterns in different activities.

2.5 Movement patterns in ACL-D patients during non-gait ADL

Observing movement patterns of ACL-D subjects during tasks other than gait may aid in determining whether the strategies used to stabilize the knee during gait are general compensation mechanisms or a task specific alterations. There have been several studies that have evaluated the movement patterns of ACL-D subjects during step up tasks, forward lunges, stepping over tasks, during destabilizing perturbations, and during stair ascent and descent. Like gait studies, there are inconsistent results regarding movement patterns of ACL-D subjects during other activities. Therefore, the remainder of this section will concentrate on the studies that have separated ACL-D subjects into Copers and Non-Copers.

2.5.1 Forward Lunges

Comparing and evaluating the performance between Copers and Non-Copers require tasks that load the knee joint. One task is that of the forward lunge where the involved limb is the limb that is loaded. Alkjaer et al. (2002) attempted to quantify the differences in knee joint movement patterns during a forward lunge between ACL-D Copers and Non-Copers. Like gait studies, Copers tended to flex their knees greater than the Non-Copers and controls. Non-Copers also experienced a significant decrease of knee extensor moment during last 25% of the lunge where knee extension is the primary action (Alkjaer et al., 2002). During the descent and ascent phases of the lunge the Non-Copers had a smaller negative peak power during the descent and smaller positive power during the ascent indicating less activation of the quadriceps muscle group while Copers and controls had similar power profile during both the descent and ascent phases (Alkjaer et al., 2002). Copers also tended to co-contract their hamstrings and quadriceps muscle groups with greater efficiency when compared to Non-Copers. This allowed them to perform the forward lunge similar to the controls.

2.5.2 Stepping over task

A study by Rudolph et al. (2004) evaluated the differences between controls, Copers, and Non-Copers during a stepping over task where one leg ascended onto the step while the other leg moves up and over the step and then landing on the floor. This study found similar results between Copers and Non-Copers as Alkjaer et al. (2003) did during gait. During the weight acceptance portion of the trial the injured leg of the Non-Copers had a decreased amount of knee flexion when compared to the Copers and controls. When the injured leg was the landing leg both the controls and Copers had similar net internal knee extensor moments while the Non-Copers had a significantly decreased net internal knee extensor moment. Rudolph et al. (1998) found similar results as this latter study between Copers and Non-Copers. Non-Copers flexed their knees less to ascend a step with their injured leg than they did with their uninjured leg.

The differences between Copers and Non-Copers during the step task are also evident during the pattern of EMG activation. The Copers group activated the lateral hamstring muscle earlier with a longer duration than the Non-Copers, while the Non-Copers displayed a delayed and decreased amount of activation in the vastus lateralis muscle. These results are similar those previously described in gait studies where Copers tend to activate the hamstring muscles to stabilize the knee while Non-Copers tended to decrease quadriceps activity rather than increase hamstring activity (Alkjaer et al., 2003). Non-Copers seem to have inadequate quadriceps muscle activity that contributes to their unsuccessful stabilization strategies (Rudolph et al., 2004). In both cases, Copers and non-Copers use co-contraction as a means to stabilize the knee. However, Copers are able to activate the quadriceps muscles without excessive anterior translation of the tibia (Chmielewski et al., 2005). Non-Copers also co-contract but they suffer from aberrant muscle activation patterns that may lead to the bouts of knee instability commonly seen in Non-Copers (Chmielewski et al., 2005). Copers and Non-Copers also exhibit different kinematic patterns during activities that lack large knee excursions. While being perturbed during unilateral stance, Non-Copers increased their knee flexion angle to provide a greater mechanical advantage to the hamstring muscles to reduce anterior tibial translation (Chmielewski et al., 2005). Copers on the other hand displayed knee angles that were between the controls and the Non-Copers.

2.5.3 Stair Ascent and descent in ACL-D subjects

2.5.3.1 In-vivo studies

Although the forward lunge and perturbations to unilateral stance are important in studying the differences in movement patterns in ACL-D subjects, they are not representative of common activities of daily living. It is important for individuals to ascend and descend stairs safely, effectively and without pain. The inability to do so can have a negative impact on persons overall quality of life and deter them from participating in a variety of activities. The repetitive nature of this activity along with the forces associated with stair ascent and descent represent a large risk for OA development. Osteoarthritis accounts for more trouble with climbing stairs and walking than any other disease (Guccione et al., 1994). Therefore, it is important to the dynamic control strategies used by normal and ACL-D subjects, both Copers and Non-Copers during stair ascent and descent to understand how ACL-D subjects load the knee joint so therapeutic measures can be implemented to prevent the development of OA. Stair ascent and descent is a challenging task, as well as an ADL. The better we are at returning an individual to normal health, the more likely we are to slow the development of OA.

The main kinematic differences in stair climbing when compared to level walking occur in the sagittal plane. During level walking the knee reaches flexion angles of around 20° while during stair climbing flexion angles increase approximately to 75° (Nadeau et al., 2003). There are fewer significant differences between walking and stair climbing in the frontal plane. For example, during the early stages of the stance phase in stair ascent the knee adducts to about 10° while during level walking the knee remains relatively stable in the frontal plane (Nadeau et al., 2003).

Normal stair climbing produces knee moments much greater than those during normal walking. Costigan et al. (2002) found that the flexion moment of the knee is more than 2 times greater in stair climbing than it is for level walking (1.16 Nm/kg compared to 0.54 Nm/kg). This increase in positive flexor moment at the knee is countered by a greater activation of the quadriceps muscle group tending to pull the tibia forward, putting strain on the ACL to resist anterior tibial translation (Costigan et al., 2002). However, during level walking most loads placed on the knee are during the first 20° of flexion. During stair climbing, the maximum load at the knee occurs around 60° of flexion which may add higher stresses and potentially more wear on the articulating surfaces (Costigan et al., 2002). The shapes of the moment profiles at the knee are consistent in the literature. Most studies indicate that during the ascent phase of stair climbing there is an initial knee extensor moment generated by the quadriceps muscles followed by a knee flexor moment. The differences lie within the values of these moments. Andriacchi (1980), Kowalk (1996), and Costigan (1993) found knee extensor moments around 60 Nm, while McFadyen & Winter (1998) found values close to 100 Nm.

ACL-D subjects reach peak knee flexion moments at approximately the same knee angle as controls (Thambyah et al., 2004). This corresponds to 16 % of the gait cycle where the trail limb foot begins to leave the floor (Thambyah et al., 2004). The peak knee flexor moment was almost 50% smaller in the injured limbs of the ACL-D when compared to the controls and the uninjured limbs (Thambyah et al., 2004). Decreased knee flexor moments during stair climbing contradict results found by Berchuck et al. (1990). Berchuck et al. (1990) concluded that although ACL-D subjects decreased knee flexor moments during level walking due to a ‘quadriceps avoidance’ strategy they did not impose the same strategy during stair climbing. Results from Berchuck et al. (1990) indicate that the knee flexor moment actually increases while ascending stairs.

Differences in the results of these two studies may lie within the step height used for stair climbing. Smaller step heights used in Thambyah et al. (2004) result in decreased knee flexor moments due to the fact that at small knee angles the quadriceps muscles employ more anterior pull on the tibia (Grood et al., 1984). Therefore, it is necessary to select a stair height that is likely to be seen in everyday life.

Results indicating a decreased knee flexor moment in ACL-D during stair climbing coincide with results during other movements (step task, forward lunge). Previous movements indicate that ACL-D may use alternate knee joint kinematics to perform activities. During stair climbing, Thambyah et al. (2004) claim that there is no difference between knee angles between ACL-D subjects and controls. ACL-D subjects did have larger knee joint excursions during stair climbing but they were not of significance.

Therefore, there is a need to describe knee joint loading patterns during stair climbing in ACL-D subjects, both Copers and Non-Copers, and couple this with EMG data from muscles surrounding the knee joint to gain insight into adaptations seen in these ACL-D subjects. Since, osteoarthritis development is believed to be related to the loads placed on the joint, it is important to understand the 'normal' loading of the knee across activities and how the loading environment is influenced by joint dysfunction or joint injury.

2.5.3.2 In-vitro studies

Taylor et al. (2004) examined the tibio-femoral joint contact forces during stair climbing using human subjects with implanted instrumented femoral prosthesis. In this study normal subjects produced considerably higher tibio-femoral contact forces during stair climbing than in walking. Ranges in peak tibio-femoral contact forces in walking have been found to be between 2.97- 3.33 times body weight (BW) while in stair climbing it has range of 5.23 – 6.16 times BW (Taylor et al., 2004). Loading of the knee during stair climbing is significantly increased, when compared to walking, when the knee is flexed greater than 15°. Antero-posterior shear forces are also drastically increased during stair climbing compared to walking. The average value of peak antero-posterior shear forces during walking is around 0.5 times BW while during stair climbing shear forces more than double to around 1.3 times BW (Taylor et al., 2004). Similarly, when the knee angle is increased past 15° in stair climbing the shear loading is increased. Therefore, during stair climbing when knee angles are similar to those of walking the loading patterns are also the same. However, during the early stages of stair ascent when knee angles are greater than 15°, tibio-femoral forces and antero-posterior shear forces are increased (Taylor et al., 2004). The shear forces at the knee are carried by the soft tissue structures of the knee, mainly by the cruciate ligaments. Any disruption to the ACL would cause the distribution of these forces to other structures of the knee and therefore cause and degenerative effect to these soft tissues (Taylor et al., 2004). This mechanism would have a greater effect during stair climbing where the shear forces are more than doubled than that of normal walking. It is then important to look at the different joint moments during stair climbing following ACL-D and identify any major differences that will negatively affect the remaining soft tissues and if ACL-D Copers and Non-Copers kinematics and kinetics differ during stair climbing as they do during walking and other movements.

Due to the fact the primary role of the ACL is to resist anterior tibial displacement, it is important to evaluate the parts of the stair climbing cycle when the tibia is displaced the most. Ahmed & McLean (2002) found a 4 – 6mm increase in anterior translation during the terminal segment of the stance phases and the beginning of the swing phase during stair climbing in a simulation using fresh-frozen knee specimens. It is likely that during the terminal segment of the stance phase and the beginning of the swing phase of stair climbing cycle differences in the dynamic control strategies between Copers and Non-Copers will become evident due to a need to prevent excessive anterior tibial translation. Ahmed and McLean (2001) provide insight into the role of the ACL with regard to tibial translation in vitro; however there is limited research into the in vivo adaptations during stair climbing in ACL-D subjects. Even more limited research is done during stair climbing that separates ACL-D subjects into Copers and Non-Copers.

2.6 Anatomical changes in ACL-D subjects

2.6.1 Muscle Volume

Functional differences between ACL-D Copers and Non-Copers may be in part due to differences in the muscle volume surrounding the knee joint. Williams et al. (2005) showed that Non-Copers had a significant amount of quadriceps atrophy when compare to Copers and a control group. However, quadriceps muscle atrophy was limited to only the vastus lateralis and vastus intermedius and not the vastus medialis (Williams et al., 2005). Quadriceps muscles were the only muscle to show a significant amount of volumetric differences between limbs (injured, uninjured) and across groups (controls, Copers, Non-Copers). The Copers group had significantly larger vastus lateralis muscles on their involved sides compared to their uninvolved sides, while the Non-Copers had significantly smaller vastus lateralis muscles on their involved sides when compared to their uninvolved sides (Williams et al., 2005). This may represent an adaptive strategy used by Copers to promote knee stabilization strategies. Stronger muscles are needed to dissipate and generate loads in the presence of knee instability in order to maintain dynamic stability of the knee (Williams et al., 2005).

2.6.2 Muscle Specificity

Williams et al. (2003, 2005) measured the ability of the muscles surrounding the knee joint to produce force in a controlled fashion. Each muscle surrounding the knee joint produces muscle activity that is related to a specific direction. The ability to coordinate all the muscles surrounding the knee joint allows us to carry out ADL. The impaired control of this function produces abnormal movements.

The specificity of principal direction of actions of the quadriceps muscle group is also significantly different between Copers and Non-Copers. Williams et al. (2003, 2005) used a specificity index to determine the specificity of action of muscles. A score of zero indicated that the muscle was active equally in all target directions while a score of 1 indicated that the muscle was active only in 1 specific direction. Overall, ACL-D subjects had a lower specificity index when compared to both their uninjured leg and the control groups (Williams et al., 2003). When partitioning out the Copers and Non-Copers, the Non-Copers showed a significantly less specificity index for the vastus lateralis muscle when compared to Copers and the control group (Williams et al., 2005). Lower specificity indeces for the vastus lateralis muscle may cause Non-Copers to fire the quadriceps in instances where it is not necessary and in fact be counterproductive (Williams et al., 2005). The Copers group displayed muscle specificity indices of the hamstrings that were lower than those of the other groups indicating that their hamstrings muscles were active across a broader range of target directions (Williams et al., 2005). The previous studies indicate a clear difference in quadriceps muscle morphology and control between Copers and Non-Copers that may impact their movement patterns.

There are quantifiable differences between Copers and Non-Copers in activities of daily living. Non-Copers tended to increase their knee angle only when the perturbation was likely to cause anterior tibial translation. Increasing the knee angle in this instance allowed the hamstring muscles to prevent the anterior tibial translation that was usually prevented by the ACL. In other motor tasks (step up, lunge), Non-Copers tended to decrease their knee angle compared to Copers and controls. This compensation is known as the ‘stiffening strategy’ (Rudolph et al., 1998). The ‘stiffening strategy’ may be the result of co-contraction which would increase the joint compression forces without actually providing stability in all conditions such as, perturbations as seen in Chmielewski et al. (2005).

Non-Copers are characterized as experiencing larger amounts of the knee giving way. Giving way of the knee joint may increase the shear forces at the knee, and combined with possible increase in joint compression forces due to co-contraction, could contribute to the degeneration of articular cartilage (Rudolph et al., 1998). Copers, on the other hand, are able to maintain joint motion that is similar to that of controls while still using co-contraction as a means of joint stability. However, maintaining joint loading patterns that is similar to controls should decrease joint compression and shear and decrease the likelihood of these forces affecting the integrity of the articular cartilage in a negative way (Rudolph et al., 1998).

2.7 Functional Differences between ACL-D Copers and Non-Copers

ACL-D Copers and Non-Copers exhibit differences in not only ADL but also in responses to surveys that discuss knee function (Knee Outcome Survey, Global Rating Scale) and during functional tests that challenge the stability of the knee joint (hop tests).

The Knee Outcome Survey (KOS) is a self-report survey that is used to determine the functional level of patients with injuries. The activities of daily living portion of this survey assesses how the patients knee condition affect different aspects of activities of daily living such as kneeling, sitting, squatting, and stair climbing. There are a total of 17 questions regarding the patient's ability to perform daily activities. The score on the KOS scale is determined by summing the point values for the responses to all 17 items on the scale. The score is then divided by 80, which is the total possible number of points for all the items on the scale; and then multiplying by 100 to get a percentage (Irrgang et al., 1998).

The KOS scale in APPENDIX 1 was developed based on existing scales that determine knee function including the Cincinnati Knee Scale, the Lysholm Knee Scale, the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC), and the guidelines developed by the International Knee Documentation Committee (IKDC) (Irrgang et al., 1998). The KOS, Lysholm Scale, and global rating scale were administered to a number of patients repeatedly (1 week, 4 weeks, 8 weeks) at nine physical therapy centres across the USA. The KOS scale was substantially higher than the Lysholm knee scale during the test-retest reliability measures. The KOS scale had coefficient alpha of 0.92-0.93 compared to the 0.60 -0.73 of the Lysholm scale reflecting a smaller standard error measurement for the KOS scale. The KOS scale also had strong correlations to scores from the other tests across the 8-week period. Compared to the Global Rating Scale, the KOS correlated with values of 0.66- 0.75 across the 8 week period. When compared to the Lysholm scale the correlations were between 0.78- 0.86. These are significant because when the Lysholm scale was correlated to the Global Rating scale values were only between 0.53- 0.54 across the 8- week period. Therefore, the KOS is a reliable, valid, and inexpensive instrument for the assessment of functional limitations that result from a wide variety of pathological disorders and impairments of the knee (Irrgang et al., 1998).

ACL-D Copers tend to have a significantly higher score on the KOS scale compared to Non-Copers (Eastlack et al., 1999). Copers tend to have an average KOS score of 95-

100% while Non-Copers tend to fall below 80%. Therefore a cut-off score of 80% was used to distinguish between Copers and Non-Copers by Fitzgerald et al. (2000). Of the 28 candidates that scored higher than 80% on the KOS scale, 79% returned to pre-injury levels without any episodes of giving way indicating that the 80% cut off is a good indicator of those who can be termed as Copers.

The Global Rating Scale is a scale that assesses the patient's knee function from a scale of 0 – 100 points, with 100 points representing pre-injury level of knee function and 0 indicating complete loss of knee function. This scale has been shown to correlate well to the KOS scale. Eastlack et al. (1999) showed that Non-Copers typically had global rating scales of less than 60% while Copers showed scores above 80%.

ACL-D Copers and Non-Copers perform differently during functional tasks that challenge the stability of the knee joint, such as hopping tasks. Detailed explanations of the hops are found in Noyes et al. (1991). The hop tests consist of 4 trials with each leg: a) a single hop for distance, b) a triple cross over hop for distance where the subject must cross over a 15 cm wide tape with each consecutive hop, c) a straight triple hop for distance, and d) a timed hop while the subject hops 6 metres. Each subject typically performs two practice trials followed by two measured trials for each leg as per Fitzgerald et al. (2000). The hops for distance are as follows:

- The single hop test is a single hop for distance with one leg;
- For the triple hop test the patients stands on one leg and performs three consecutive hops as far as possible and landing on the same foot;
- The cross over hop is performed on a course with a 15 cm marking strip on the floor which extends approximately 6 metres. The subject is required to hop

three consecutive times on one foot, crossing over the centre strip on each hop. The total distance of the hop will be measured. (Noyes et al., 1991)

The hops measured for distance are expressed as a percentage of the injured leg divided by the uninjured leg. However, the timed hop score are expressed as a percentage of the uninjured extremity score divided by the injured extremity score (Fitzgerald et al., 2000).

The reliability of this test was studied by Bolgla et al. (1997) and found the correlation of the scores for the distance hops to be between 0.95- 0.96 when performed on consecutive days by the same individuals with standard errors of 4.56, 15.44, and 15.95 cm for the single hop, triple hop, and the cross over hop. The timed 6 metre hop had lower correlation of 0.66. This study shows that functional tasks, such as these hop tests, are reliable when measuring the function of the lower extremity. Due to the fact that isokinetic quadriceps strength is not strongly correlated with performance during functional tasks, these tasks need to be performed to assess the performance and function of the lower limb. Greenberger et al. (1995) found that these hop test correlated to quadriceps strength by values of only 0.78 for the dominant leg and 0.65 for the non-dominant leg. Copers have been discriminated from Non-Copers if they obtain values greater than 90% for the timed hop as suggested by Fitzgerald et al. (2000) and scores greater than 80% for the distance hops as per Eastlack et al. (1999).

2.8 Marker Placement

Manal et al. (2000) examined eleven different marker arrays over the tibia to assess the best marker set to track tibial motion. The marker arrays differed by either being constrained by a thermoplastic shell or unconstrained, on the lateral part of the shank or medial border of the tibia, or by being under-wrapped or over-wrapped. The eleven different marker arrays were compared to the data measured from a percutaneous skeletal tracker (PST) anchored on the medial and lateral malleoli. The marker set that best resembled the data measured with the PST was markers placed over the distal, lateral aspect of the tibia. Therefore, that marker array was be used in this study. This method of attachment ranked as the best method for tracking tibial motion (Manal et al. 2000).

2.9 Conclusions

Using kinematic, kinetic, and electromyographical measures that will be described later, information and insight will be provided into the dynamic control strategies used by volunteers with ACL-D knees to stabilize the knee joint during stair climbing. Previous research has shown that ACL-D subjects show differences in movement patterns and can be defined as belonging two 2 distinct groups; Copers and Non-Copers. Information regarding differences in movement patterns between groups is necessary in establishing knee joint loading patterns, especially during stair climbing so that a better understanding of these alterations can be established and their resulting affect on the development of OA. Rehabilitation methods geared towards maintenance of normal knee joint loading patterns may be developed in an attempt to decrease risk of OA development.

As a result of this study differences between ACL-Deficient Copers and Non-Copers were identified during a more challenging ADL. The differences between groups could be used to identify potential factors that may increase the likelihood of one group developing additional joint pathologies, such as osteoarthritis. These differences can be utilized by the rehabilitation field by providing specific rehabilitation protocols to ACL-Deficient Copers and Non-Copers based on their different strategies used to achieve and accomplish normal ADL.

3. Methods

3.1 Subjects

Twenty subjects volunteered for this study. Ten subjects were recruited from the university community (5 male, 5 female). These subjects had healthy knees and regularly participated in physical activity at least 3-5 times per week. The remaining 10 subjects (7 females, 3 males) represented the ACL-D groups of the study. Five of these subjects represented the 'Copers' group and the other 5 represented the 'Non-Copers' group. Copers included those who had returned to pre-activity level with no surgery to repair the ACL. Non-copers represented those who had not returned to pre-injury level and experienced several bouts of knee instability. Each of these ACL-D subjects had a unilateral rupture of the ACL with no other concomitant knee injuries such as additional knee ligament injury, fracture or dislocation in the knee, full thickness cartilage defects, knee joint effusion, or other lower extremity pathologies at the time of their participation.

The ACL-D subjects were recruited from the University community with referral by the Head Athletic Therapist at the University of Waterloo and from Sports Medicine Clinics and Orthopaedic Surgeons from the Kitchener-Waterloo area. The average time post ACL injury was approximately 1 year for the patient groups.

ACL rupture was confirmed by an orthropaedic surgeon arthroscopically, MRI or by other standard clinical tests such as knee joint laxity test. All ACL-D subjects had an uninvolved knee that was healthy and had full range of motion in both the injured and uninjured knees. Subjects gave informed consent on a form approved by the Office of Research Ethics at the University of Waterloo.

3.2 Screening Examination

Each ACL-D subject came into the lab before the stair climbing tasks to be screened in order to determine which subjects were ACL Copers and Non-Copers. The screening examination consisted of the following tests: a) the Knee Outcome Survey (KOS) Activities of Daily Living Scale (Irrgang et al. 1998), b) Global Rating of Knee Function, and c) self report episodes of knee giving way. The screening examination protocol, including the surveys, are included in Appendix 1.

The algorithm used to select a Coper was:

- Greater than 80% on the KOS activities of daily living score; and
- Global rating of function greater than 60% of pre-injury level (Fitzgerald et al., 2000); and

Less than two episodes of giving way with activities of daily living since injury as per Rudolph et al. (2004).

3.3 Protocol

3.3.1 Preliminary Screening

The screening examination was the only part of the procedure that differed between the ACL-D and control groups. The ACL-D subjects reported to the lab for an hour prior to the start of the stair climbing task. The subjects read and signed the informed consent form approved by the Office of Research Ethics at the University of Waterloo. The ACL-D subjects then filled out the KOS- Activities of Daily Living Survey and the Global Rating of Function Score.

After the ACL-D subjects finished the screening examination their scores were tabulated. Based on the results of the scores they were identified as either an ACL Coper or Non-Coper for the remainder of the testing. The participants were not notified if they were a Coper or Non-Coper during the testing. The remainder of the testing procedure was the same for both the control group and the ACL-D groups. All subjects were in the laboratory for approximately 2 – 3 hours.

3.3.2 Task

EMG electrodes were placed on the subjects as described in section 3.6. The subjects then performed maximum voluntary exertions (MVE's) for the lower limb muscles where EMG electrodes were placed. Subjects then stood quietly and a 10s resting EMG trial was collected and assumed to be baseline EMG values. OptoTrak markers were placed on the anatomical landmarks and on the segments of the lower limb to track 3-D motion of the lower limb as described in section 3.4.

The subjects participated in 20 trials of stair climbing (10 ascent and 10 descent) on stairs (4 steps) with a rise of 20 cm and a run of 30 cm. The dimensions of the stairs were chosen based upon the standards of the Ontario Building Code. The Ontario Building Code states that maximum allowable rise of stairs should not exceed 20 cm and the maximum allowable run should not exceed 35.5 cm. Therefore, we chose a rise of 20 cm assuming this will provide the greatest knee flexion. The run was chosen as a midpoint of the minimum and maximum allowable run based on the Ontario Building Code. The subjects were able to perform as many practice trials as necessary to become accustomed to the stairs and to get used to the additional EMG and OptoTrak leads. When the participants felt they were able to perform the stair climbing task as normally as possible the collection process began. For each ACL-D subject, 10 randomized trials were led with the injured leg (5 ascent and 5 descent) and 10 randomized trials (5 ascent and 5 descent) were led with the uninjured leg so that comparisons could be made between groups and between injured and uninjured legs. For the control subjects, 10 randomized trials were performed with the right leg (5 ascent and 5 descent) as the lead leg and the left leg as the trail limb.

Then 10 randomized trials were completed with the left leg as the lead limb (5 ascent and 5 descent) and the right as the trail limb.

3.4 Kinematics

The kinematic data was obtained using a twelve camera active three-dimensional motion capture system (Optotrak Certus, Northern Digital Inc. Waterloo, Ontario). Kinematic data was sampled at a frequency of 64 Hz to coincide with force plate data. All kinematic data was low pass filtered using a dual pass 4th order Butterworth filter with a cut off frequency of 6 Hz.

Active infra-red emitting diodes were placed bilaterally over the right and left anterior superior iliac spine (ASIS), medial and lateral femoral condyles, head of fibula, tibial tuberosity, medial and lateral malleolus, lateral calcaneus, and the fifth metatarsal. These markers were used to create a calibration trial of the individual so that joint centres could be defined. The locations of the joint centres are outlined in Table 1:

Table 1: Location of Joint Centres

Joint	Joint Centre
Ankle	Midway point between the lateral malleolus and medial malleolus markers.
Knee	Midway between the medial and lateral femoral condyles.
Hip	Based on the distances between the ASIS markers, the hip joint centre was defined as -21%, -32%, and -34% for the x, y, and z axes, respectively, from the corresponding ASIS marker (Shea et al., 1997).

The femoral anatomical coordinate system was defined starting at the knee joint centre. The y-axis was a line passing through the knee joint centre to the hip joint centre (positive being superior). The z-axis passed along a plane defined by the medial and lateral femoral condyles and orthogonal to the y-axis (positive going from left to right), and the x-axis acted orthogonal to the y-z plane (positive being anterior) (Besier et al., 2003).

Thermoplastic shells with four markers rigidly attached were placed bilaterally on the distal lateral thigh and distal lateral aspect of the shank. These markers were used to track the motion of the lower limb and knee. It has been previously shown that tibial rotation is better estimated when the marker arrays are placed more distal than proximal over the lateral shank (Manal et al., 2000).

An under-wrapped method was used to secure the thermoplastic plates to the thigh and leg segments. The under-wrap attachment consisted of wrapping an elasticized band around the part of the segment where the plates were attached. The thermoplastic shells were fitted with a Velcro backing that attached to the elasticized band.

3.5 Kinetics

Ground reaction forces (GRF) were measured with a force platform (AMTI OR6-7, Advanced Mechanical Technology Inc, Watertown, MA). The calibrated force platform was fitted into Step 2 of the custom built staircase. Steps 1 and 3 were mounted around the force platform as per Thambyah et al. (2004). The GRFs were sampled at a rate of 2048 Hz and later down sampled and synchronized with the kinematic data. The natural frequency of the force platform is 370 Hz in the x and y directions and 530 Hz in the z direction, which is well above any GRF frequency that will be collected in this study.

Three-dimensional net joint internal moments of force were calculated about the ankle and knee joints using an inverse dynamics approach. Moments were normalized to each subject's body mass. After joint moments were normalized, peaks in joint moments were identified and the different peaks and % stance to peaks were investigated between groups.

3.6 Electromyography

Bipolar (Ag-AgCl) surface electrodes were applied to 14 muscles with an inter-electrode distance of 2 cm. EMG signals were collected using 2 AMT-8 EMG Wire Telemetry systems (Bortec Biomedical Ltd., Calgary, AB) with a frequency response of 10-1000 Hz, an input impedance of 10 GΩ and a common mode rejection ratio of 115 dB at 60 Hz. The electrodes were placed bilaterally over the muscle bellies of the muscles shown in Table 2. The placements of the electrodes were according to the guidelines by Delagi et al. (1981), and are described in Table 2.

Table 2: Placement of EMG Electrodes

Tibialis Anterior	Four fingerbreadths below the tibial tuberosity and one fingerbreadth lateral to the tibial crest.
Medial Gastrocnemius	One hand breadth below the popliteal crease on the medial mass of the calf
Lateral Gastrocnemius	One hand breadth below the popliteal crease on the lateral mass of the calf.
Semitendinosus	Midway on a line between the medial condyle of the femur and ischial tuberosity
Biceps Femoris	Midpoint of a line between the head of the fibula and the ischial tuberosity.
Vastus Lateralis	Over the lateral aspect of the thigh, one handbreadth above the patella.
Vastus Medialis	Four fingerbreadths proximal to the superiomedial angle of the patella.

EMG electrodes were placed over skin that had been shaved, lightly abraded, and cleaned with isopropyl alcohol. After the EMG electrodes were placed on the muscles, maximum voluntary exertions (MVE) were collected for each muscle. These MVE were used to normalize EMG data collected during the stair climbing task. Three MVEs were collected for each of the muscles with a period of rest (approximately 2 minutes) between trials to minimize fatigue. The raw EMG data for each muscle was full wave rectified (FWR) and low pass filtered (as described later). The greatest value recorded of the three consecutive trials was considered the MVE (De Luca, 1997). MVE measurement for each muscle is described in Table 3.

All EMG data was recorded at a sampling rate of 2048 Hz. The EMG signals were then full wave rectified and low pass filtered with a single pass 2nd order Butterworth filter with a cutoff frequency of 10 Hz to create a linear envelope EMG. Peak activity was selected after the EMG data had been filtered and linear enveloped.

Table 3: Methods of collecting maximum voluntary exertions (MVE)

Tibialis Anterior	Subjects were seated with knee extended to approximately 100°. Subjects actively dorsiflex the ankle against resistance.
Medial and Lateral Gastrocnemius	Subjects seated upright with hip, knee, and ankle flexed at 90°. Subjects plantar flex with resistance on the knee to prevent heel elevation (Todd et al., 2004).
Semitendinosus	Subjects were prone on the table with hip in neutral position and knee flexed at 90° and were asked to flex at the knee while resistance was applied in the opposite direction on the shank (Onishi et al., 2002).
Biceps Femoris	Subjects were prone on the table with hip in neutral position and knee flexed at 60° and were asked to flex at the knee while resistance was applied in the opposite direction on the shank (Onishi et al., 2002).
Vastus Lateralis and Medialis	Knees were extended at 15° knee flexion and hip in neutral position, the subjects were asked to extend their knee while resistance was applied in the opposite direction on the shank (Mirzabeigi et al., 1999).

3.7 Data Analysis

The data analysis consisted of comparisons between three groups; Control group (left limb), ACL-D Coper group (injured limb), and ACL-D Non-Coper group (injured limb). The left limb was chosen in the control group to make data processing in the frontal plane simpler.

For each phase of stair climbing (ascent and descent), each participant's trials were normalized to 100% of stance. Using linear interpolation, the data samples were increased or decreased evenly across the stance cycle to equal 101 data points. These trials were then ensemble averaged for each participant to a single trial which was used in the statistical analysis. Table 4 provides a list of variables that were extracted from the data collected. The variables were compared across groups in an attempt to identify any significant differences between the groups. All variables were compared using a one-way analysis of variance (ANOVA) with a post hoc HSD test using a significance level of 0.05. The variables were compared across groups for both the ascent and descent phases of the stair climbing task.

Table 4: List of Variables Calculated

Kinematic	Knee angle at onset
	Knee angle at 25% stance
	Knee angle at 50% stance
	Knee angle at 75% stance
	Knee angle at toe-off
Kinetic	Knee Flex-Ext Moment – Early: 0 – 35% of stance (N-m/Kg)
	Knee Flex-Ext Moment – Mid: 35 – 80% of stance (N-m/Kg)
	Knee Flex-Ext Moment – Late: 80 – 100% of stance (N-m/Kg)
	Knee Abd-Add Moment – Early: 0 – 40% of stance (N-m/Kg)
	Knee Med-Lat Moment – Late: 70 – 95% of stance (N-m/Kg)
	Ankle Flex-Ext Moment – Early: 0- 35% of stance (N-m/Kg)
	Ankle Flex-Ext Moment – Mid: 35 – 70% of stance (N-m/Kg)
	Ankle Flex-Ext Moment – Late: 70 – 100% of stance (N-m/Kg)
	Knee Flex-Ext Power – K1 (W/Kg)
	Knee Flex-Ext Power – K2 (W/Kg)
	Knee Flex-Ext Power – K3(W/Kg)
	Knee Flex-Ext Power – K4 (W/Kg)

	Ankle Flex-Ext Power – A1 (W/Kg)
	Ankle Flex-Ext Power – A2 (W/Kg)
	Ankle Flex-Ext Power – A3 (W/Kg)
	Ankle Flex-Ext Power – A4 (W/Kg)
Electromyography	Peak EMG Values
	Occurrence of Peak EMG Values
	Average EMG
	Cross Correlation of Vastus Medialis/ Biceps Femoris
	Cross Correlation of Vastus Lateralis/ Semitendinosus

The magnitudes of muscle activation (AEMG) during the stance phase were determined for the vastus lateralis, vastus medialis, biceps femoris, semitendinosus, and medial gastrocnemius by integrating the linear envelope curve from the initiation of force plate contact to the point of toe off and then dividing by the stance time. In order to quantify the level of co-contraction, cross correlation analysis was performed on two sets of muscles. The values obtained during the cross correlation analysis were at a time lag of zero. The r values obtained were then compared across groups.

4. Results

A total of 20 participants were collected including 10 (5 male and 5 female) Healthy controls, 5 (2 male and 3 female) ACL-D Copers, and 5 (1 male and 4 females) ACL-D Non-Copers. The Healthy group had an average weight of 75.9 kg ±16.1 kg and an average height of 174.3 cm ±12.1cm, while the Copers group had an average weight of 71.7 kg ±6.2 kg and an average height of 170.8 cm ±5.6 cm, and the Non-Copers group had an average weight of 71.3 kg ±13.1 kg and an average height of 168.7 cm ±4.8 cm. Table 5 provides descriptive information of the injury status' ACL-Deficient groups. In each of the ACL-D groups there were 3 subjects who had injured their left knee and 2 subjects that injured their right knee. The most common test that was failed among the Non-Coper group was the number of occurrences where the injured knee gave way, where 4 out of the 5 Non-Copers had equal to or greater than 2 occurrences. The only Non-Coper that did not have equal to or greater than 2 occurrences where their knee gave way had a KOS score of 60 which placed them in the Non-Copers group.

Table 5: ACL Deficient Injury Status Information (Items that are bolded indicate the tests that the ACL-D subject failed that led them to be placed in the Non-Coper group)

Group	Gender	Injured Limb	Months Since Injury	# of 'Giving Way' Occurrences Since Injury	Global Rating of Function	KOS Score
Coper	Female	Left	4	1	85	87.5
Coper	Male	Right	9	0	85	95
Coper	Male	Right	2	1	80	80
Coper	Female	Left	26	0	80	83.75
Coper	Female	Left	2	1	75	88.75
Non Coper	Female	Right	3	**2**	90	92.5
Non Coper	Male	Left	5	**2**	90	91.25
Non Coper	Female	Left	49	**4**	60	**68.75**
Non Coper	Female	Left	3	0	70	**60**
Non Coper	Female	Right	22	**4**	85	**77.5**

4.1 Ascent

4.1.1 Electromyography

Electromyographical data was collected on 7 different muscles on the limb during the stance phase. Peak EMG, percent to stance at which peak EMG occurred, and average EMG (AEMG) were collected for each muscle and compared across groups. The results are shown in Table 6, Figure 1 ,Figure 2 , Figure 3 and Ensemble averaged EMG profiles for each muscle are also shown in Figure 4 through Figure 10.

The Healthy group was able to achieve higher peak EMG values with respect to maximum voluntary exertion for each of the measured muscles except for the Biceps Femoris where the Non-Copers group achieved the highest value. The Non-Copers group achieved higher peak EMG values over the Copers group for the vastus medialis, medial gastrocnemius, biceps femoris, and semitendinosus, while the Copers group achieved higher peak EMG values over the Non-Copers for the lateral gastrocnemius, vastus lateralis, and the tibialis anterior. The muscles that had the largest peak EMG values were both the gastrocnemius muscles and the vastus medialis and vastus lateralis muscles where the peak EMG values were approximately 50 – 70 % MVE. The semitendinosus and biceps femoris muscles did not achieve large peak EMG values with measured values around 20% MVE. The tibialis anterior was also relatively quiet with peak EMG values of between 20 – 30% MVE. There were only two instances where there were significant differences in peak EMG between groups. For both the vastus medialis and medial gastrocnemius, the Healthy group achieved higher peak EMG values when compared to the Copers group (p-value= 0.04 and p-value= 0.01, respectively).

Table 6: EMG Results – Ascent (* indicates areas of significant differences)

	Copers			Non-Copers			Healthy		
	Peak EMG (SD)	Percent Stance to Peak (SD)	AEMG (SD)	Peak EMG (SD)	Percent Stance to Peak (SD)	AEMG (SD)	Peak EMG (SD)	Percent Stance to Peak (SD)	AEMG (SD)
Medial Gastrconemius	*54.2 (9.2)**	*85 (4.2)*	*17.0 (5.2)*	*60.1 (7.2)*	*67 (29.3)*	*18.9 (4.6)*	*72.6 (8.5)**	*82 (3.6)*	*16.5 (3.0)*
Lateral Gastrocnemius	51.2 (20.7)	87 (33.8)	14.4(4.9)	49.4 (17.4)	72 (33.8)	15.8 (6.8)	70.7 (10.6)	84 (3.0)	12.2 (4.5)
Biceps Femoris	19.0 (23.9)	87 (28.6)	4.7 (4.8)	19.4 (13.8)	82 (17.2)	9.5 (7.1)	16.4 (6.2)	57 (38.2)	4.3 (2.0)
Semitendinosus	14.9 (11.7)	75 (11.7)	3.6 (3.3)	19.9 (13.1)	56 (45.6)	7.5 (5.6)	21.4 (6.8)	59 (40.8)	4.4 (3.6)
Vastus Lateralis	48.1 (13.1)	17 (5.1)	13.9 (2.4)	43.9 (24.9)	30 (30.0)	12.7 (6.4)	58.4 (15.9)	15 (3.8)	17.7 (6.5)
Vastus Medialis	*47.8 (13.8)**	*18 (4.1)*	*19.1 (14.7)*	*59.6 (9.9)*	*31 (29.2)*	*16.4 (4.2)*	*67.7 (11.9)**	*14 (3.2)*	*15.9 (4.9)*
Tibialis Anterior	23.3 (13.1)	34 (35.4)	9.3 (4.9)	19.2 (2.2)	12 (11.9)	9.1 (1.2)	32.4 (11.3)	33 (34.8)	8.0 (3.2)

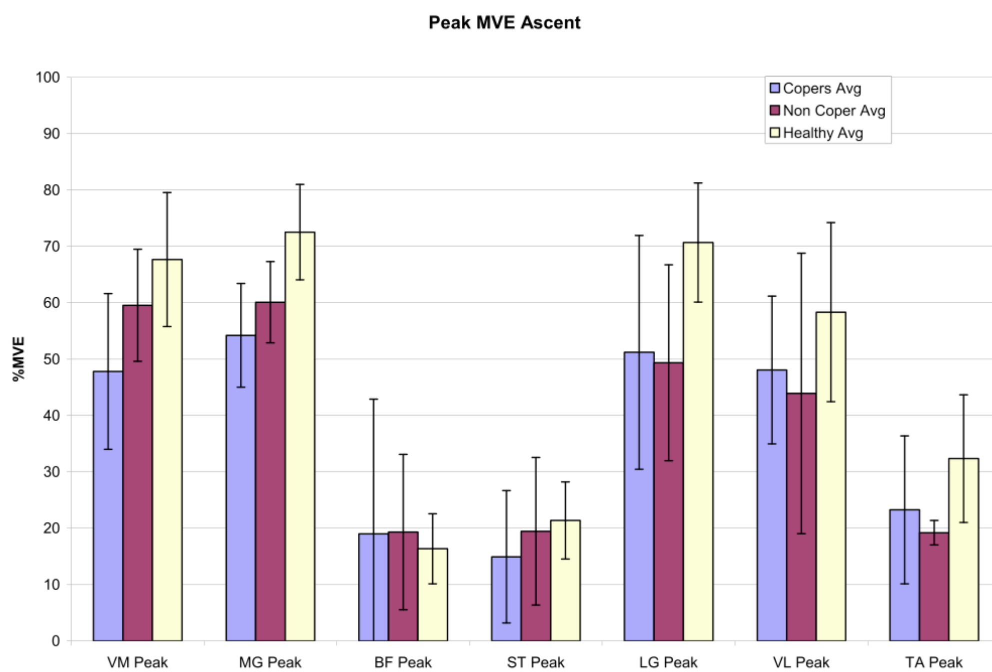

Figure 1: Peak MVE Values during Stair Ascent

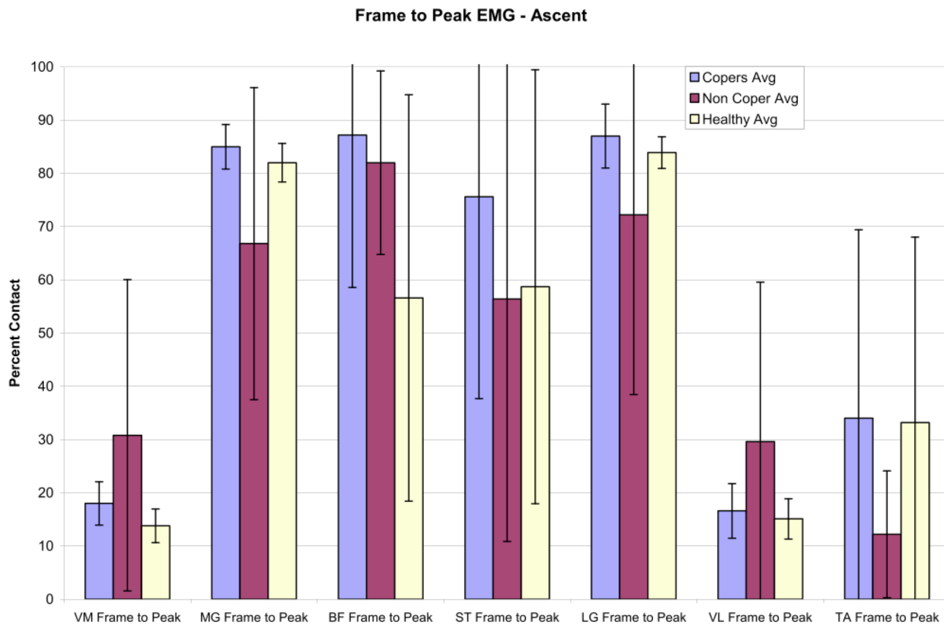

Figure 2: Percent to Peak of Muscles during Stair Ascent

The percent of the stance phase when these muscles achieved their maximum %MVE was also measured and is graphically displayed in Figure 8. The muscles that achieved their maximum activity early on in the stance phase were both the vastus medialis and vastus lateralis muscles which provided the initial burst to propel the body up the step. The tibialis anterior muscle also achieved its peak activity relatively early on in the stance phase. The Healthy and Copers group achieved peak EMG activity of the vastus medialis and vastus lateralis muscles at approximately 15% of stance phase while the Non-Copers group peak was delayed until approximately 30% of the stance phase. Due to the large variability between subjects within the Non-Copers group these differences were not found to be statistically significant. With respect to the tibialis anterior the Non-Copers group actually achieved peak EMG earlier than the Healthy and Copers group at around 12% stance while the Healthy and Copers group achieved peak tibialis anterior values at 33 – 34% of stance. Both the Healthy and Copers group had peak gastrocnemius values at approximately 85% of the stance phase for both the medial and lateral gastrocnemius. However, the Non-Copers group activated their medial gastrocnemius and lateral gastrocnemius earlier on in the stance phase at 60% and 72% of the stance phase respectively. These differences between groups were not found to be statistically significant. The biceps femoris muscle also reached it peak activity in the last half of the stance phase. The Healthy group reached its peak biceps femoris value at 57% of the stance phase while both the Copers and Non-Copers group reached their peak biceps femoris value at 87% and 82% of the stance phase respectively. Due to the large variability within each of the groups these differences were not found to be significant. Both the Healthy group and the Non-Copers group reached peak semitendinosus EMG values at approximately 60% of the stance phase while the Copers group reached peak semitendinosus activity at 75% of the stance. Due to the large variability within each of the groups these differences were found to be not significant.

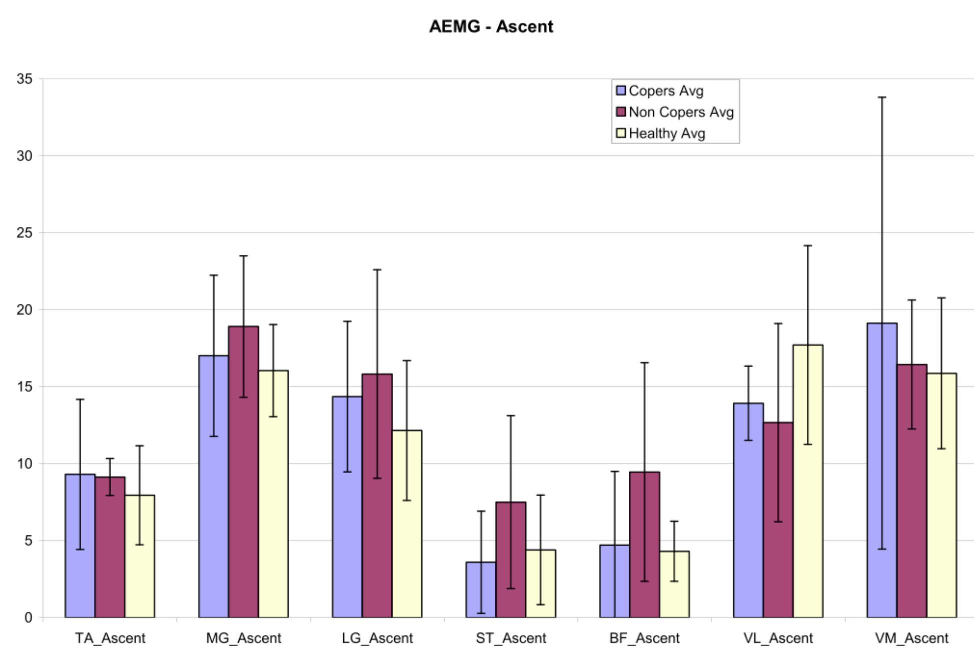

Figure 3: AEMG Values during Stair Ascent

Average EMG (AEMG) was calculated for each of muscles of the stance leg during the stance phase. The values are shown in Figure 9 with the units in %MVE. The most active muscles were the medial and lateral gastrocnemius and the vastus medialis and the vastus lateralis with the muscles using between 12 – 18% MVE. The least active muscles were the semitendinosus, biceps femoris, and the tibialis anterior with values below 10% MVE.

No significant differences were observed between the groups for any of the muscles. However, the Non-Copers group tended to use the hamstrings and gastrocnemius muscle group more than the other two groups, while the Copers used a greater amount of vastus medialis muscle and the Healthy used a greater amount of the vastus lateralis muscle.

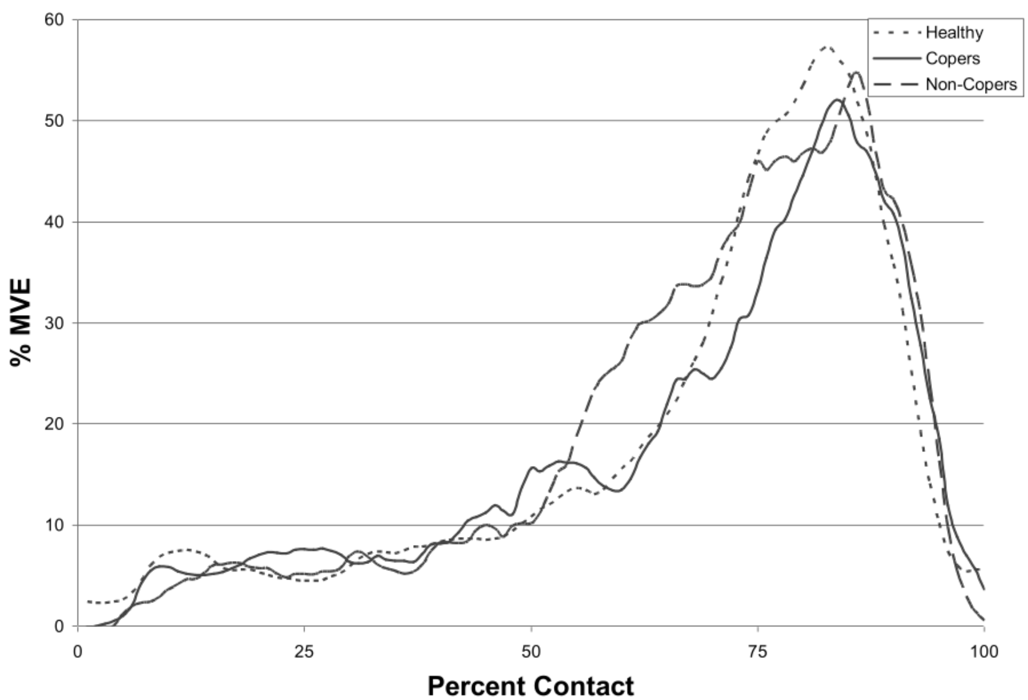

Figure 4: Medial Gastrocnemius Profiles during Stair Ascent

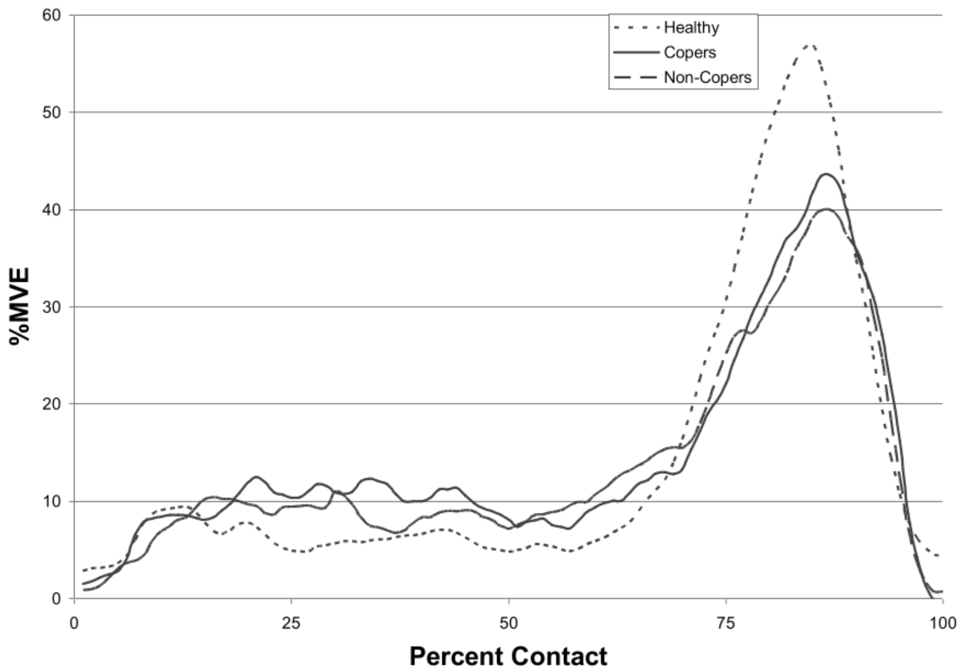

Figure 5: Lateral Gastrocnemius Profiles during Stair Ascent

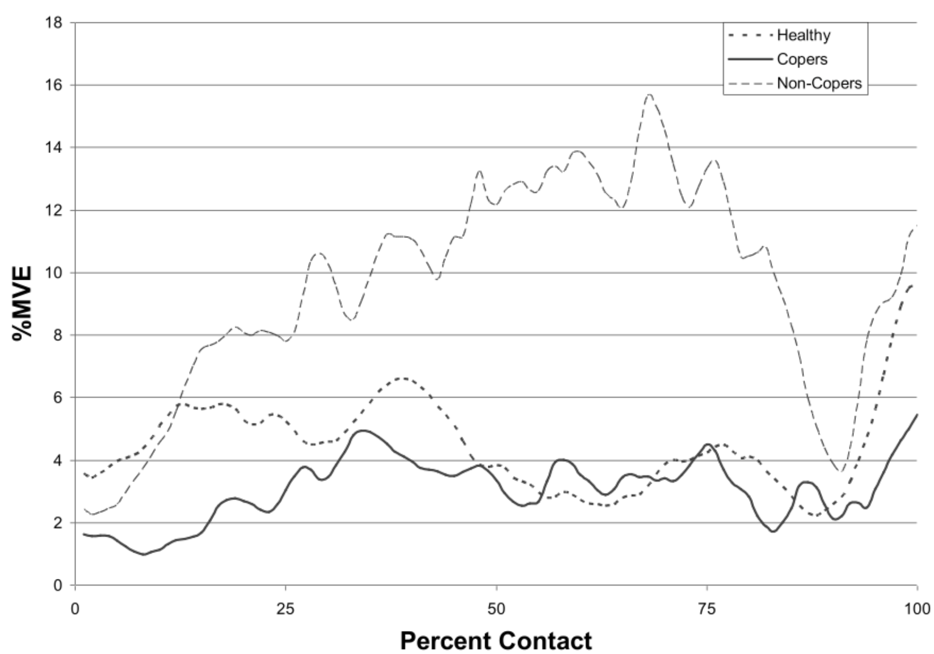

Figure 6: Biceps Femoris Profiles during Stair Ascent

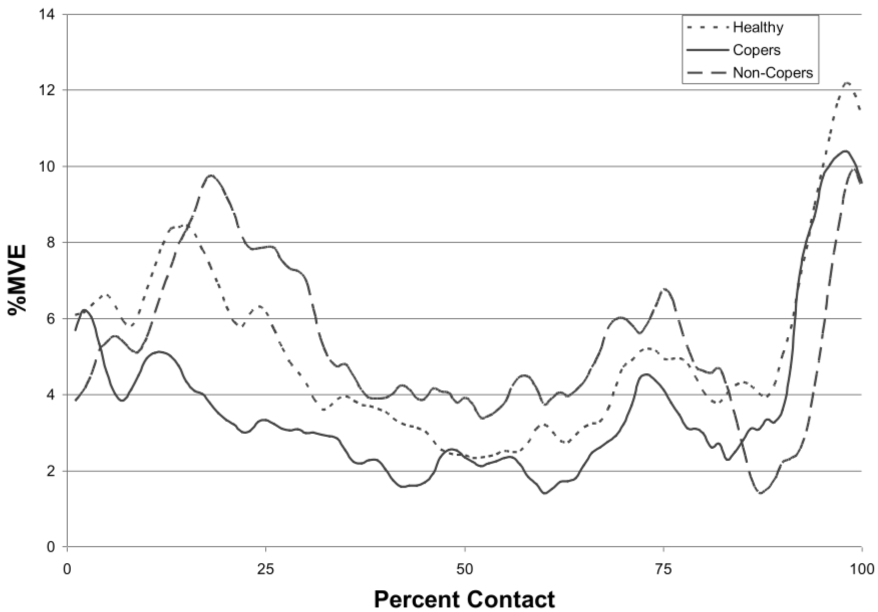

Figure 7: Semitendinosus Profiles during Stair Ascent

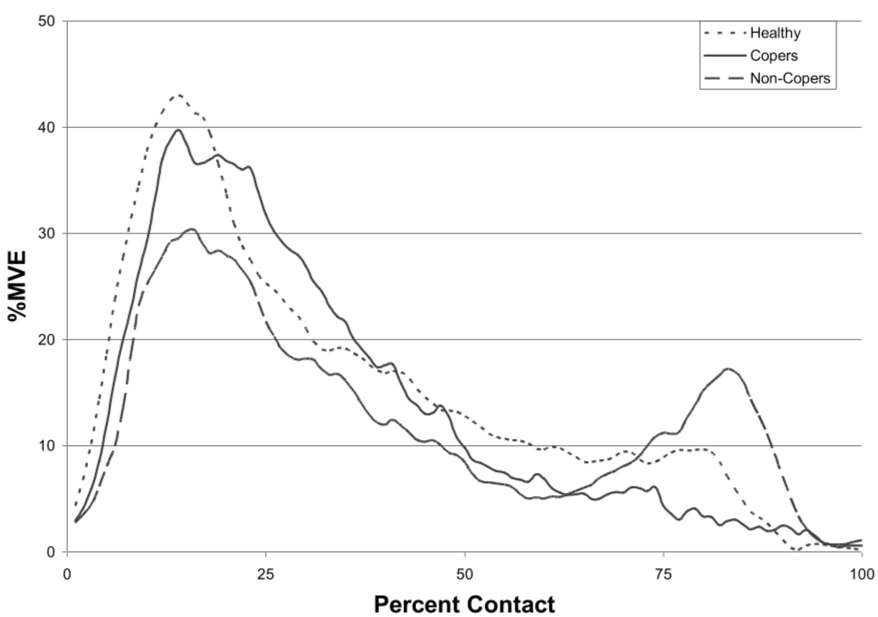

Figure 8: Vastus Lateralis Profiles during Stair Ascent

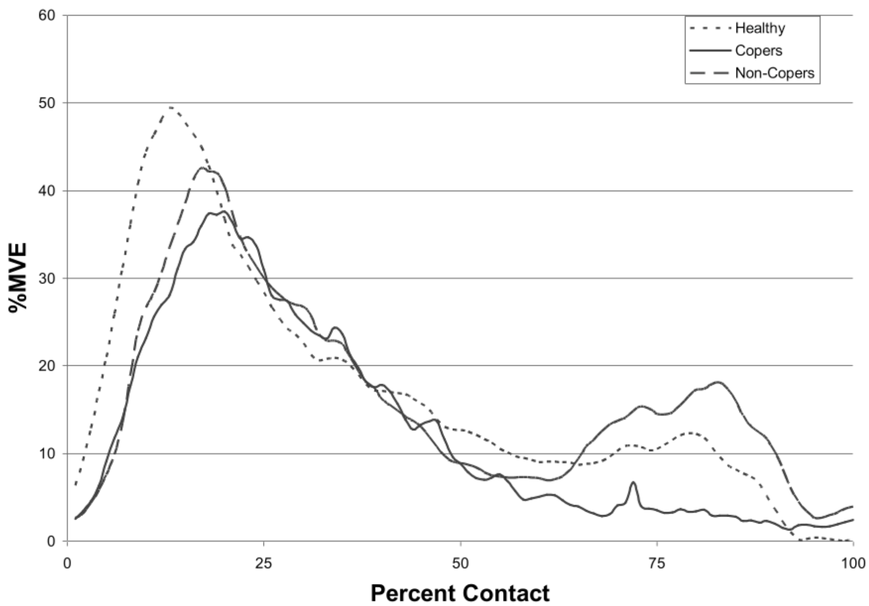

Figure 9: Vastus Medialis Profiles during Stair Ascent

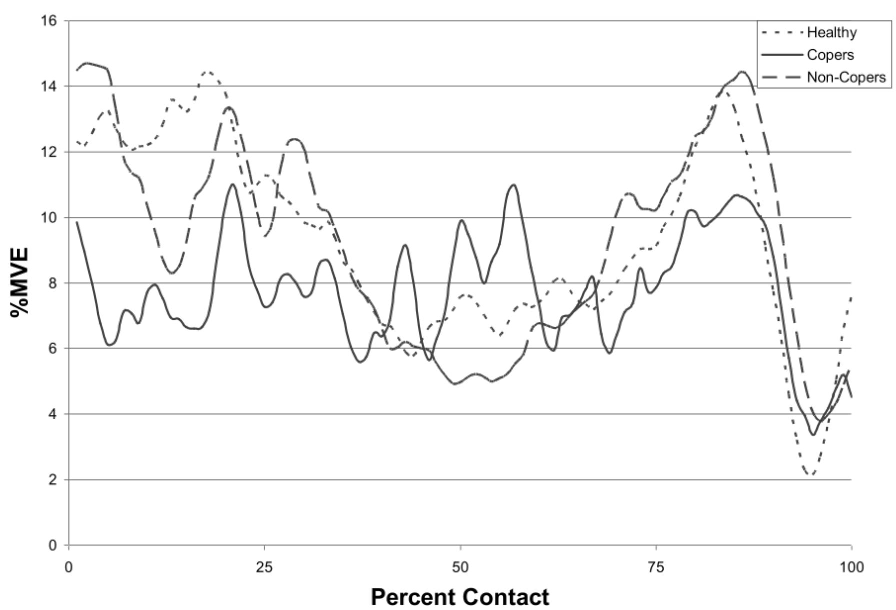

Figure 10: Tibialis Anterior Profiles during Stair Ascent

4.1.1.1 Cross Correlation

Cross correlation analyses were performed on two sets of muscles representing the medial and lateral muscle groups acting around the knee. The results of the analyses are shown in Table 7. The results refer to the correlation between the muscles at a zero time lag in an attempt to indicate the level of co-contraction on the medial and lateral sides of the knee joint. With respect to the medial side of the knee (vastus medialis/ biceps femoris), the Healthy group displayed the largest positive correlation between the two muscles compared to the Non-Copers group. The Copers group had a negative relationship between the muscles indicating that as the activity of one muscle increased the other one decreased. The differences between the groups were not found to be significantly different. With respect to the lateral side of the knee (vastus lateralis/ semitendinosus) all groups had an average positive correlation with the Non-Copers having the largest correlation followed by the Healthy group and then the Copers. None of the differences between the groups were found to be significantly different.

Table 7: Cross Correlation Results during Stair Ascent

	Copers (SD)	Non-Copers (SD)	Healthy (SD)
Vastus Medialis/ Biceps Femoris	-0.206 (0.24)	0.102 (0.42)	0.254 (0.37)
Vastus Lateralis/ Semitendinosus	0.087 (0.40)	0.217 (0.29)	0.115 (0.42)

4.1.2 Kinetics

Kinetic measures were taken during stair ascent, from the start of force plate contact until the end of contact. Knee and ankle net joint moment of force profiles were divided into three phases of the stance phase (early, mid, and late) based upon the shape of the profiles. Based on the profile for ankle power it was divided into four phases consisting of an a1, a2, a3, and an a4 phase. The profile for knee power was divided into four phases consisting of a k1, k2, k3, and a k4 phase.

Maximum and minimum values for each group and each phase are provided in Table 8, with the movement profiles shown in Figure 11, Figure 12, Figure 14, Figure 13, and Figure 15.

Table 8: Kinetics Results – Ascent (* indicates where significant differences exist)

	Copers (SD)	Copers Range	Non-Copers (SD)	Non-Copers Range	Healthy (SD)	Healthy Range
Knee Flex-Ext Moment – Early (N-M/Kg)	0.19 (0.30)	(-0.13)- 0.61	0.26 (0.17)	0.09 – 0.45	0.33 (0.15)	0.02 – 0.53
Knee Flex-Ext Moment – Mid (N-M/Kg)	-0.67 (0.31)	(-0.26) – (-0.91)	-0.52 (0.32)	(-0.17) – (-0.94)	-0.84 (0.24)	(-0.42) – (-1.24)
Knee Flex-Ext Moment – Late (N-M/Kg)	-0.14 (0.07)	(-0.21) – 0.09	0.10 (0.06)*	0.02 – 0.19	-0.21 (0.13)*	(-0.06) – (-0.47)
Knee Add-Abd Moment – Early (N-M/Kg)	0.49 (0.08)	0.39 – 0.57	0.42 (0.10)	0.32 – 0.58	0.52 (0.16)	0.33 – 0.81
Knee Add-Abd Moment – Late (N-M/Kg)	0.60 (0.13)	0.44 – 0.70	0.55 (0.08)	0.44 – 0.64	0.70 (0.17)	0.54 – 1.12
Ankle Moment – Early (N-M/Kg)	-1.98 (0.16)	(-2.21) – (-1.81)	-1.99 (0.15)	(-2.15) – (-1.77)	-2.09 (0.38)	(-2.87) – (-1.52)
Ankle Moment – Mid (N-M/Kg)	-1.53 (0.14)	(-1.72) – (-1.37)	-1.51 (0.15)	(-1.77) – (-1.39)	-1.44 (0.24)	(-1.94) – (-1.16)
Ankle Moment – Late (N-M/Kg)	-2.16 (0.17)	(-2.42) – (-1.96)	-2.07 (0.11)	(-2.18) – (-1.94)	-2.35 (0.39)	(-2.93) – (-1.55)
Knee Power – K1 (W/Kg)	0.21 (0.46)	(-0.22) – 0.82	-0.35 (0.43)	(-0.07) – 0.90	-0.33 (0.26)	(-0.79) – 0.01
Knee Power – K2 (W/Kg)	0.97 (0.27)	0.69 – 1.24	0.88 (0.58)	0.14 – 1.46	1.32 (0.41)	0.86 – 2.25
Knee Power – K3(W/Kg)	0.01 (0.11)	(-0.09) - 0.15	-0.07 (0.11)	(-0.21) – 0.07	0.01 (0.23)	(-0.27) – 0.57
Knee Power – K4 (W/Kg)	-0.05 (0.07)	(-0.04) - 0.15	0.01 (0.10)	(-0.15) – 0.12	-0.35 (0.45)	(-0.12) – 1.39
Ankle Power – A1 (W/Kg)	-0.25 (0.27)	(-0.72) – (-0.02)	-0.34 (0.13)	(-0.55) – (-0.22)	-0.70 (0.49)	(-1.75) – (-0.28)
Ankle Power – A2 (W/Kg)	1.16 (0.32)	0.63 – 1.38	1.09 (0.40)	0.65 – 1.47	1.41 (0.59)	0.69 – 2.56
Ankle Power – A3 (W/Kg)	-0.29 (0.31)	(0.56) – 0.17	-0.17 (0.34)	(-0.61) – 0.23	-0.33 (0.30)	(-0.74) – 0.16
Ankle Power – A4 (W/Kg)	4.40 (0.96)	2.95 – 5.49	4.19 (0.58)	3.21 – 4.74	5.39 (1.59)	2.15 – 8.43

4.1.2.1 Net Joint Moments of Force

Knee Flexion-Extension Moments

The average knee flexion-extension moment for each percent of the stance phase was calculated for each group and 3 distinct phases of the moment profile were compared between groups (early, mid, and late). The knee flexion-extension moment profiles are shown in Figure 11. The profiles illustrate that the Copers and Non-Copers group achieved consistent net flexor moments during the mid stance phase that were consistently lower than the Healthy group. The Healthy group also achieved larger net extensor moments during the early phase of stance over the two condition groups. An important observation is that the Non-Copers group achieved a net extensor moment during the late phase of stance while both the Healthy and Copers group remained in a net flexor moment. Each group contacted the force plate with a net flexor moment and the proceeded into a net extensor moment. During the early phase the Healthy group achieved the largest flexor moment followed by the Non-Copers group, and then by the Copers group. The difference between the groups during the early phase was not found to be statistically different. The middle phase represented a large sustained net flexor moment with the Healthy group achieving the largest flexor moment followed by the Copers group and then by the Non-Copers group. The differences between groups during mid stance were not found to be statistically different. The late phase represented the Healthy and Copers groups decreasing there net flexor moments to values of -0.206 N-M/Kg and -0.140 N-M/Kg respectively. The Non-Copers group actually achieved a net extensor moment during the last 10% of the stance phase and achieved a peak extensor moment of 0.098 N-M/Kg. The Non-Copers group and the Healthy group were found to have significantly different knee moment values during the late phase of stance (p-value = <0.01. The differences between the Healthy group and the Copers and the difference between the Copers and Non-Copers were not found to be statistically significant.

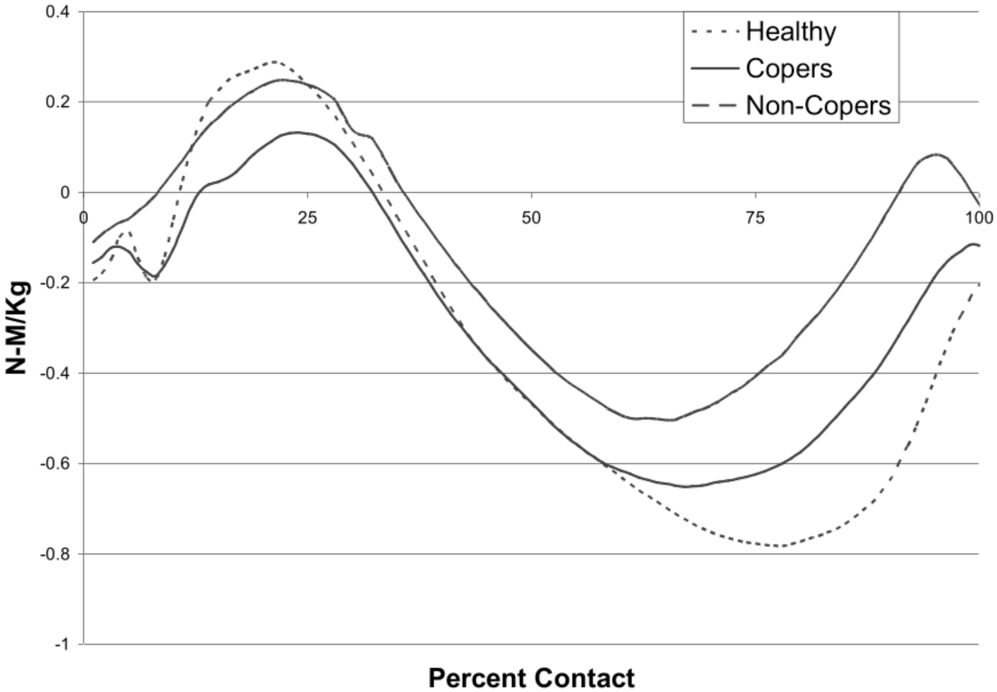

Figure 11: Knee Flexion-Extension Moment Profiles during Stair Ascent (positive values refer to extensor moments and negative values refer to flexor moments)

Knee Adduction-Abduction Moments

The average knee adduction-abduction moment for each percent of the stance phase was calculated for each group and 2 distinct phases of the moment profile were compared between groups (early and late). The knee adduction-abduction moment profiles are shown in Figure 12. All groups maintained an abduction moment on the knee throughout the entire stance phase. The adduction-abduction moment profile is characterized by 2 large peaks separated by a decrease in abduction moment. At the onset of contact onto the force plate all groups had approximately the same level of abduction moment. Throughout the majority of the remaining portion of the stance phase the Healthy group maintained a larger abduction moment followed by the Copers group and then the Non-Copers group. At the point at which the stance phase ended, the Copers and Healthy group had very similar abduction moments while the Non-Copers had a slightly lower adduction moment. As previously mentioned, during the first peak the Healthy group had the largest abduction moment with a value of followed by the Copers group, and then by the Non-Copers group. The differences between the groups during the early peak were not found to be statistically significant. During the late peak the Healthy group achieved the largest abduction moment, then the Copers group, and then the Non-Copers. The differences between the groups during the late peak were not found to be statistically significant.

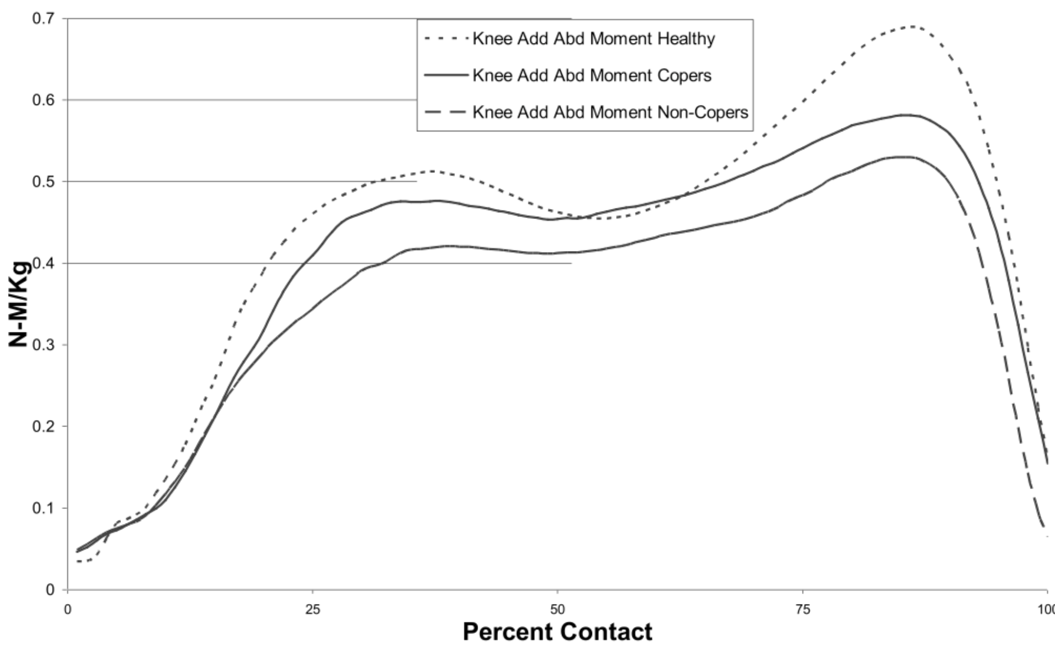

Figure 12: Knee Abduction Moments during Stair Ascent

Ankle Flexion-Extension Moment

The average ankle moment for each percent of the stance phase was calculated for each group and 3 distinct phases of the moment profile were compared between groups (early, mid, and late). The ankle moment profiles are shown in Figure 13. The profiles illustrate that the all groups achieved consistent net plantar flexor moments during all phases of stance. The Healthy group achieved larger peaks during the early and late portions of stance and smaller values during the mid portion of stance. During the early phase the Healthy group achieved the largest plantar flexor moment, followed by the Non-Copers group, and then by the Copers group. The difference between the groups during the early phase was not found to be statistically different. The middle phase represented a decreased net plantar flexor moment with the Healthy group achieving the smallest plantar flexor moment, followed by the Non-Copers group and then by the Copers group. The differences between groups during mid stance were not found to be statistically different. During the late phase all groups increased their net plantar flexor moments with the Healthy group achieving the highest value followed by the Copers group and finally then Non-Copers group. The differences between the groups were not found to be statistically different.

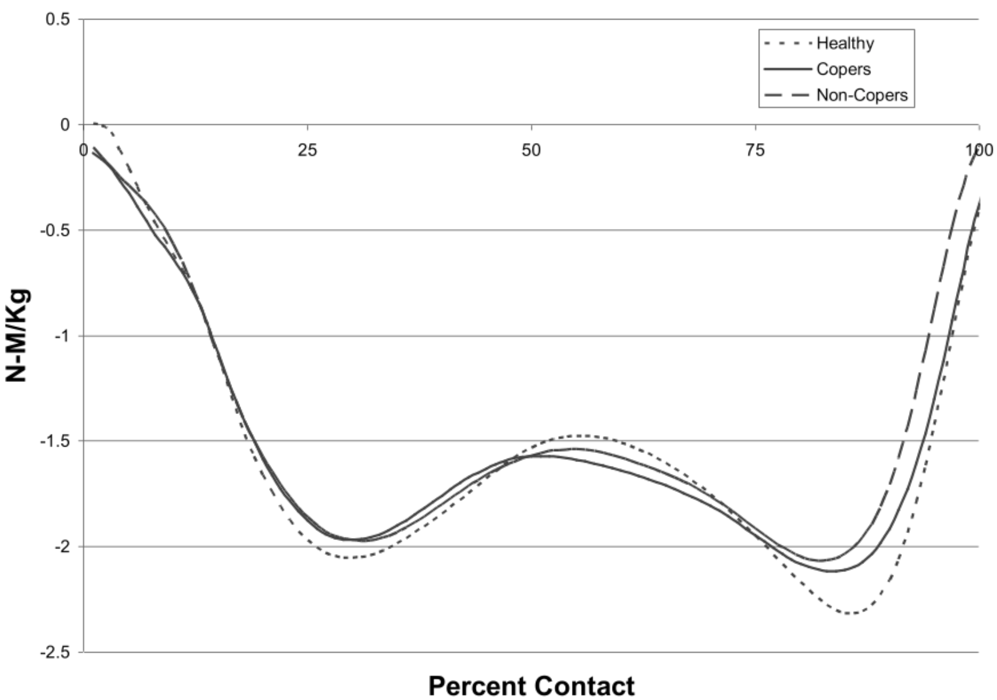

Figure 13: Ankle Moment Profiles during Stair Ascent (positive values refer to dorsi flexor moments while negative values refer to plantar flexor moments)

4.1.2.2 Powers

Knee Powers

The average knee power for each percent of the stance phase was calculated for each group and 4 distinct phases of the power profile were compared between groups (k1, k2, k3 and k4). The knee power profiles are shown in Figure 14. The profiles illustrate a period of eccentric activity (power absorption by the knee extensors) by the Healthy and Non-Copers group during k1 stance with the Copers group exhibiting concentric activity (power generation). The k1 phase consisted of the onset of stepping onto the force plate to 30% of the stance phase. Each group contacted the force plate with concentric activity of the knee flexors and only the Healthy and Non-Copers group proceeded into eccentric activity of the knee extensors . During the k1 phase the Non-Copers group achieved the largest amount of eccentric activity, followed by the Healthy group and then by the Copers group. The difference between the groups during the k1 phase was not found to be statistically different. The k2 phase represented a large sustained amount of concentric activity by the knee flexors with the Healthy group achieving the largest amount of concentric activity followed by the Copers group then the Non-Copers. The difference between the groups during the k2 phase was not found to be statistically different. The k3 phase, from 70% to 80% stance, had a decreasing amount of concentric activity for all groups with the Non-Copers group switching to eccentric activity. Maximum values during this phase illustrated that the Non-Copers had a mean eccentric maximum value of -0.066 W/Kg followed by the Copers group with a value of -0.003 W/Kg and then by the Healthy group with a value of -0.010 W/Kg. The k4 phase of stance was from 80% to 100% of the stance phase. Both the Copers and Healthy group maintained eccentric activity and had minimum power values of -0.051 W/Kg and -0.348 W/ Kg respectively. The Non-Copers group maintained concentric activity with mean minimum power of 0.012 W/Kg. The differences between the groups in each of the k3 and k4 phases were not found to be statistically different.

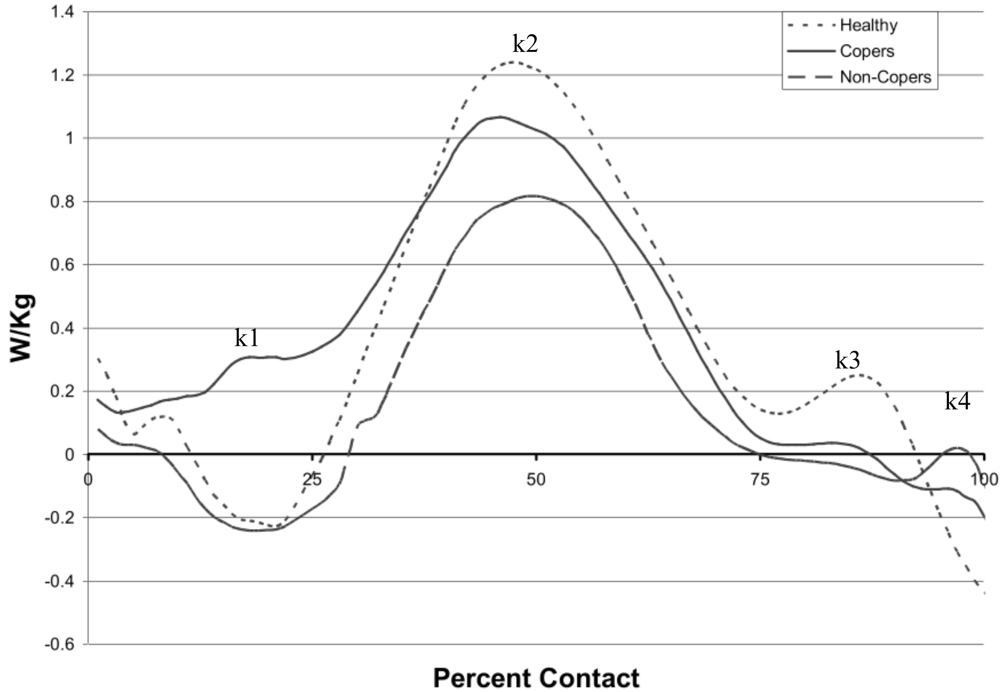

Figure 14: Knee Power Profiles during Stair Ascent (positive values indicate power generation and negative values indicate power absorption)

Ankle Powers

The average ankle power for each percent of the stance phase was calculated for each group and 4 distinct phases of the power profile was compared between groups (a1, a2 and a3, and a4). The ankle power profiles are shown in Figure 15. The profiles illustrate a period of eccentric activity in a1 followed by a sustained period of concentric activity during the a2, and then a brief instance of eccentric activity during the a3 phase and lastly a large amount of concentric activity during the a4 phase. The a1 phase consisted of the onset of stepping onto the force plate to approximately 20% of the stance phase. Each group contacted the force plate with eccentric activity of the plantarflexors. The Healthy group achieved the largest peak eccentric activity followed by the Non-Copers and then the Copers. The a2 phase took place between 20% and 60%

of the stance phase with the Healthy group achieving the highest amount of plantarflexor concentric activity followed by the Copers and Non-Copers, respectively. The a3 phase was found during 60% and 80% of the stance phase. This period featured a brief period of plantarflexor eccentric activity with the Healthy group having the largest amount of eccentric activity, followed by the Copers and Non-Copers, respectively. The a4 phase of stance featured a large amount of concentric activity with the Healthy group having the largest amount of concentric, followed by the Copers and Non-Copers, respectively. In each phase of the ankle power profile, the Healthy group had the highest amount of activity, whether it was concentric or eccentric. The a1 phase of eccentric activity was the only phase where the Non-Copers had a greater amount of concentric or eccentric activity over the Copers group. No statistical difference was found between any of the groups during any of the ankle power phases.

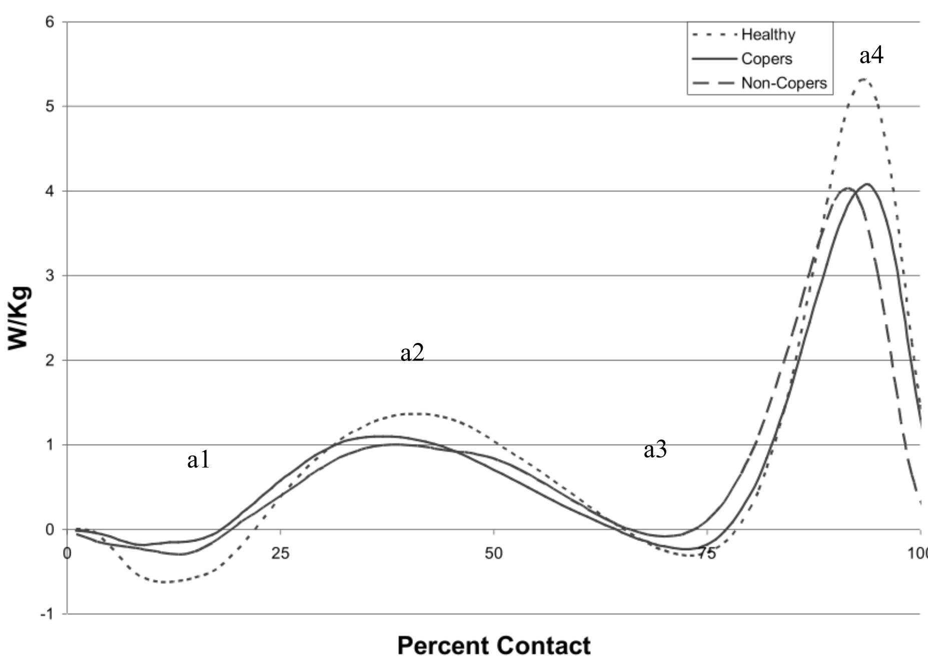

Figure 15: Ankle Power Profiles during Stair Ascent (positive values indicate power generation and negative values indicate power absorption)

4.1.3 Kinematics

Knee angles were measured for each percentage of stance and the profiles are shown in Figure 16. All groups contacted the force plate with their knee in a flexed position and during the progression to the next step their knee extended until the last part of the stance phase were there was a slight increase in knee flexion prior to toe-off. Five instances were selected for comparison across groups (onset, 25% stance, 50% stance, 75% stance, and offset) and the results are shown in Table 9. In each instance, the Non-Copers group was approximately 4 – 6 degrees more flexed than the Copers group and anywhere from 2 – 11 degrees more flexed than the Healthy group. The Copers group had greater knee flexion than the Healthy group during stance. The biggest difference observed between the groups was seen at toe-off where the Non-Copers knees were flexed on average 7 degrees more than the Copers group and 11 degrees more than the Healthy group. Even though differences were observed between the groups none of them were found to be statistically significant at any of the key instances selected.

Table 9: Knee Angles during Stance Phase of Stair Ascent

	Copers		Non-Copers		Healthy	
	Peak (SD)	Range	Peak (SD)	Range	Peak (SD)	Range
Onset – (degrees)	75.8 (7.2)	63.2 – 80.1	78.7 (5.9)	72.4 – 85.5	73.1 (6.3)	63.6 – 87.8
25% Stance – (degrees)	58.4 (6.6)	32.5 – 66.9	61.3 (5.5)	51.8 – 65.0	53.1 (13.5)	48.1 – 68.1
50% Stance – (degrees)	28.4 (5.6)	19.2 – 40.6	32.0(11.4)	18.1 – 43.8	25.3 (8.9)	21.8 – 37.7
75% Stance – (degrees)	18.0 (7.6)	10.3 – 32.8	22.2 (11.6)	11.8 – 35.6	15.7 (9.6)	6.7 – 28.8
Offset – (degrees)	22.1 (4.9)	13.9 – 31.0	28.2 (9.2)	18.3 – 39.8	20.8 (7.2)	10.6 – 26.0

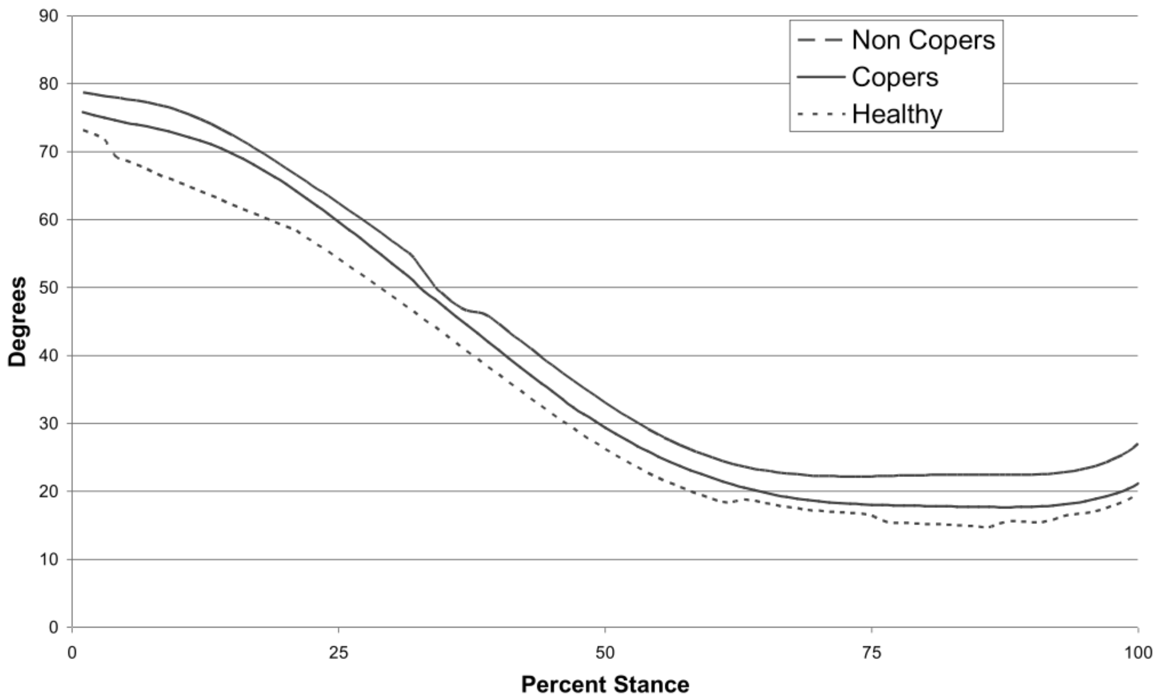

Figure 16: Knee Angle Profiles during Stair Ascent (full extension at 0 degrees)

4.2 Descent

4.2.1 Electromyography

Electromyographical data was collected on 7 different muscles during stance of the descent phase. Peak EMG, percent to stance at which peak EMG occurred, and AEMG was collected for each muscle and compared across groups. The results are shown in Table 10, and in Figure 17, Figure 18, and Figure 19. Ensemble averaged EMG profiles for each muscle are shown in Figure 20 thru Figure 26.

The Healthy group was able to achieve the highest peak EMG values with respect to maximum voluntary exertion for the tibialis anterior, vastus lateralis, and medial gastrocnemius. The Copers achieved highest peak EMG values for the vastus medialis, semitendinosus, and the biceps femoris, while the Non-Copers had the highest peak EMG value for the lateral gastrocnemius. The muscles that had the largest peak EMG values for the three groups were the vastus medialis and the medial gastrocnemius where the peak EMG values were approximately 35 - 40 % MVE. The semitendinosus, biceps femoris, and the lateral gastrocnemius muscles had the lowest peak EMG values that were below 25% MVE. The differences between the groups for all the peak EMG values were not statistically significant.

Table 10: EMG Results – Descent

	Copers			Non-Copers			Healthy		
	Peak EMG (SD)	Percent Stance to Peak (SD)	AEMG (SD)	Peak EMG (SD)	Percent Stance to Peak (SD)	AEMG (SD)	Peak EMG (SD)	Percent Stance to Peak (SD)	AEMG (SD)
Medial Gastrocnemius (MG)	30.66 (7.9)	5.60 (3.0)	6.56 (4.3)	30.06 (14.7)	21.80 (31.4)	6.02 (3.2)	42.93 (13.7)	8.40 (2.5)	5.40 (2.1)
Lateral Gastrocnemius (LG)	11.36 (2.7)	39.00 (29.4)	5.42 (2.5)	21.06 (17.1)	46.80 (32.6)	7.69 (5.9)	15.77 (5.2)	31.50 (35.5)	4.51 (2.3)
Biceps Femoris (BF)	20.02 (16.2)	78.80 (36.3)	3.30 (3.6)	19.92 (10.9)	40.60 (49.3)	5.94 (3.7)	9.72 (5.1)	54.20 (45.3)	2.22 (1.6)
Semitendinosus (ST)	20.73 (15.9)	71.80 (40.2)	3.12 (3.4)	13.24 (5.2)	62.20 (46.7)	4.80 (2.6)	12.98 (7.0)	67.20 (44.3)	1.98 (1.7)
Vastus Lateralis (VL)	29.20 (4.5)	78.40 (4.2)	11.67 (1.9)	22.75 (9.8)	55.60 (33.6)	8.65 (3.8)	38.11 (12.2)	64.20 (24.1)	11.81 (4.9)
Vastus Medialis (VM)	37.84 (9.1)	78.80 (5.0)	18.80 (13.9)	34.20 (9.7)	67.80 (31.3)	11.52 (3.7)	36.60 (14.2)	69.60 (21.9)	10.75 (5.1)
Tibialis Anterior (TA)	23.32 (6.7)	74.60 (32.9)	9.38 (3.4)	23.51 (4.7)	65.20 (44.1)	9.03 (2.1)	32.37 (7.4)	62.00 (38.1)	9.58 (1.9)

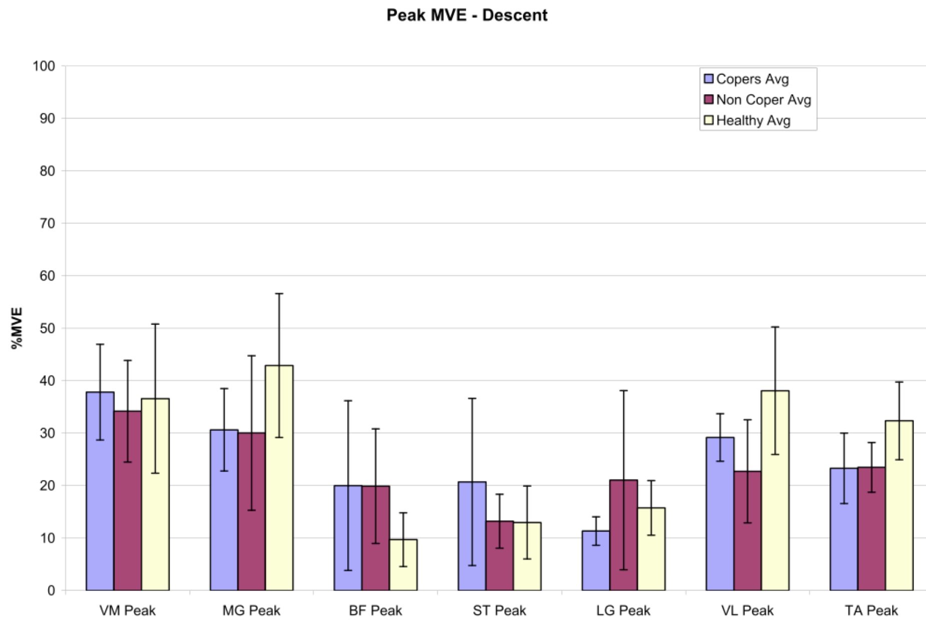

Figure 17: Peak MVE Values during Stair Descent (VM= vastus medialis, MG= medial gastrocnemius, BF= biceps femoris, ST= semitendinosus, LG= lateral gastrocnemius, VL = vastus lateralis, TA= tibialis anterior)

The percent of the stance phase when these muscles achieved their maximum %MVE was also measured and is graphically displayed in Figure 24. The muscle that achieved maximum activity early on in the stance phase for all the groups was the medial gastrocnemius which achieved maximum activity during the first 10% of stance phase for the Copers and Healthy groups and at 22% of the stance phase for the Non-Copers. For the Copers and Healthy groups, the lateral gastrocnemius was the next muscle to achieve maximal activation between 30 – 40% of the stance phase while the Non-Copers maximally activated the biceps femoris muscle next at 40% of the stance phase with the lateral gastrocnemius achieving maximal activation at 47% of the stance phase. For the hamstrings and quadriceps muscle groups, the Non-Copers group maximally activated these muscles earlier on in the stance phase compared to the other groups at approximately 40 – 70% of the stance phase. The Healthy group maximally activated the hamstrings and quadriceps muscle groups at approximately 55 – 70% of the stance phase, while the

Copers maximally activated these muscle groups between 70 – 80% of the stance phase. The tibialis anterior muscle achieved its peak activity during the 2nd half of the stance phase at 62% of the stance phase for the Healthy group, 65% for the Non-Copers group, and 75% for the Copers group. Even though there were slight differences in the timings to which peak EMG activity occurred between groups there were no significant differences found for any of the muscles between any of the groups.

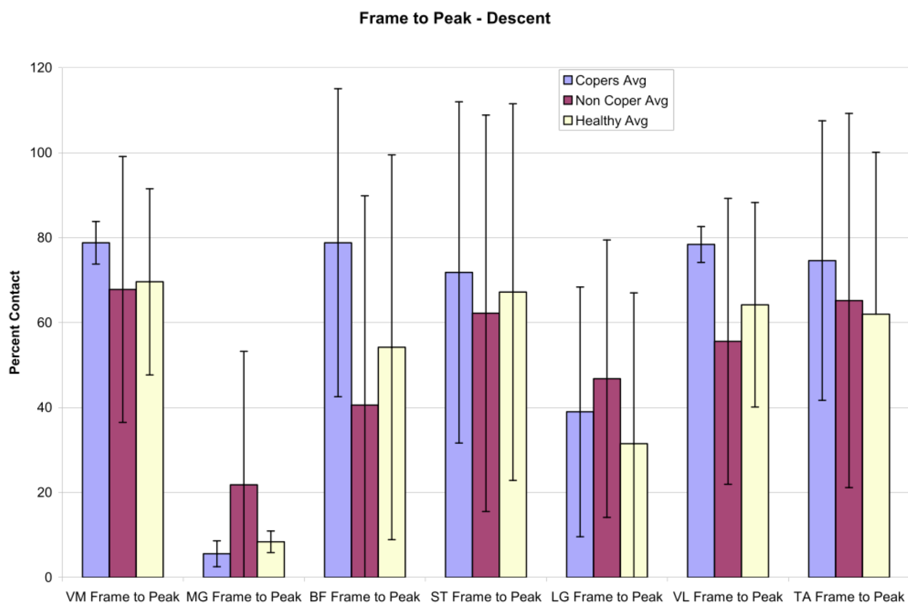

Figure 18: Occurrence of Peak EMG Values as a Percentage of Stance (VM= vastus medialis, MG= medial gastrocnemius, BF= biceps femoris, ST= semitendinosus, LG= lateral gastrocnemius, VL = vastus lateralis, TA= tibialis anterior)

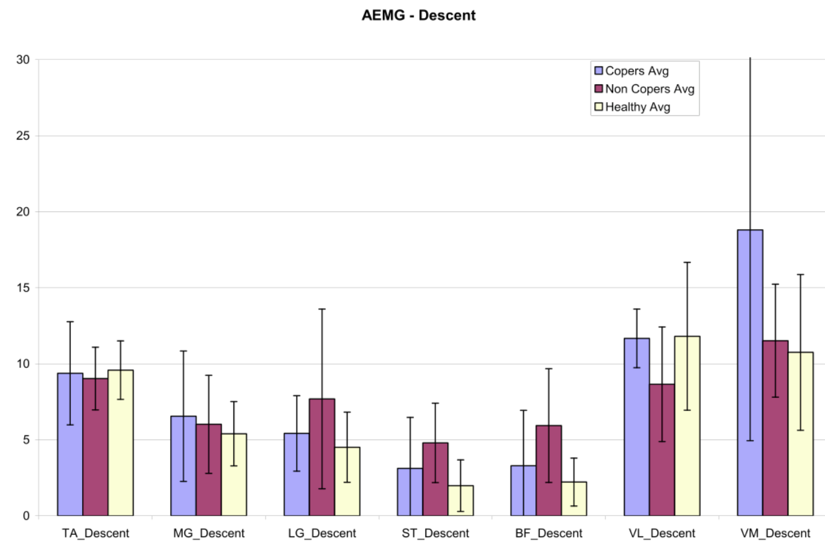

Figure 19: AEMG Values during Stair Descent (VM= vastus medialis, MG= medial gastrocnemius, BF= biceps femoris, ST= semitendinosus, LG= lateral gastrocnemius, VL = vastus lateralis, TA= tibialis anterior)

Average EMG (AEMG) was calculated for each of muscles of the stance leg during the stance phase. The values are shown in Table 10 with the units in %MVE. The muscles were less active during descent than they were during ascent. No significant differences were observed between the groups for any of the muscles.

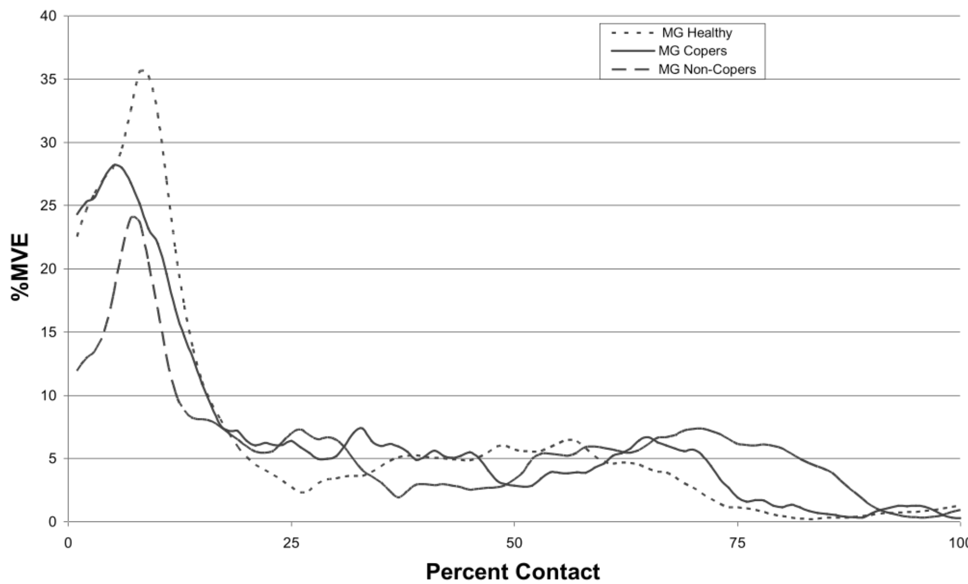

Figure 20: Medial Gastrocnemius Profiles during Stair Descent

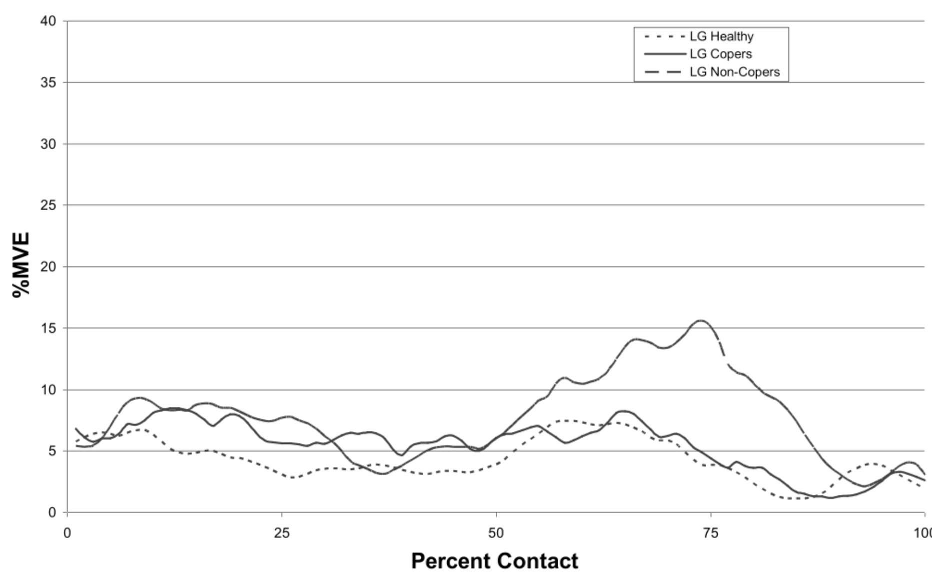

Figure 21: Lateral Gastrocnemius Profiles during Stair Descent

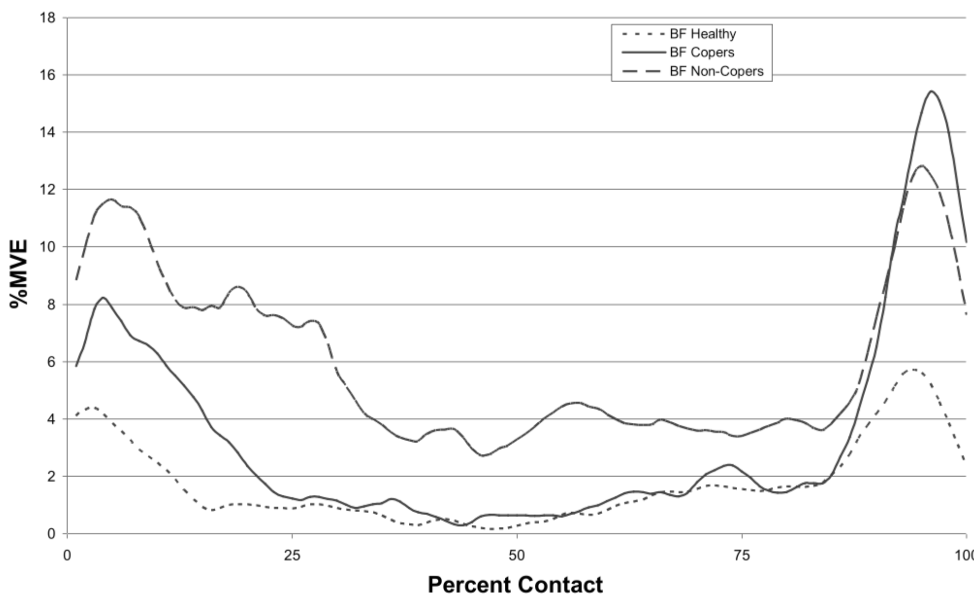

Figure 22: Biceps Femoris Profiles during Stair Descent

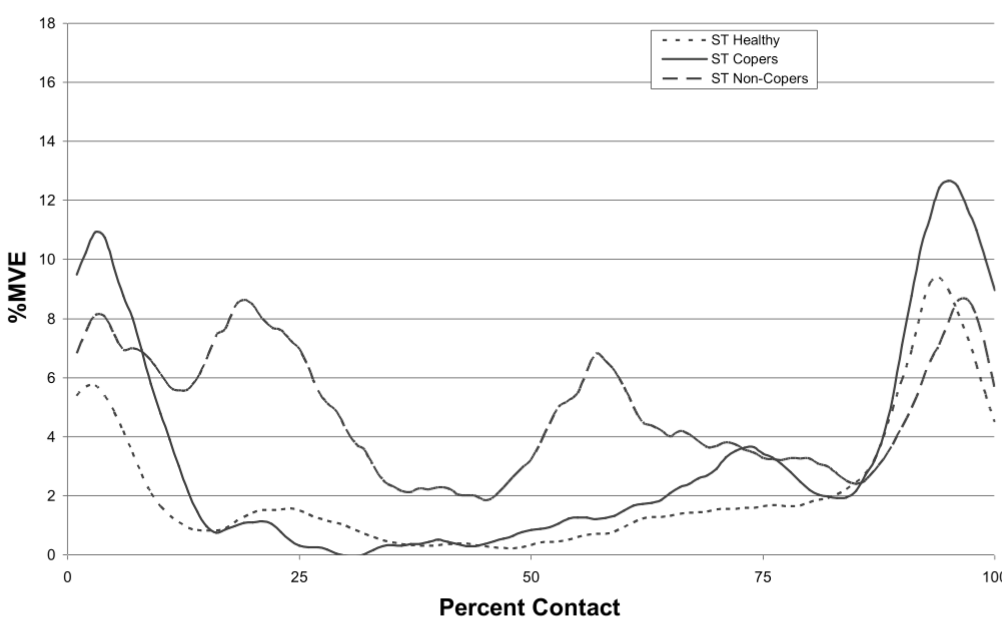

Figure 23: Semitendinosus Profiles during Stair Descent

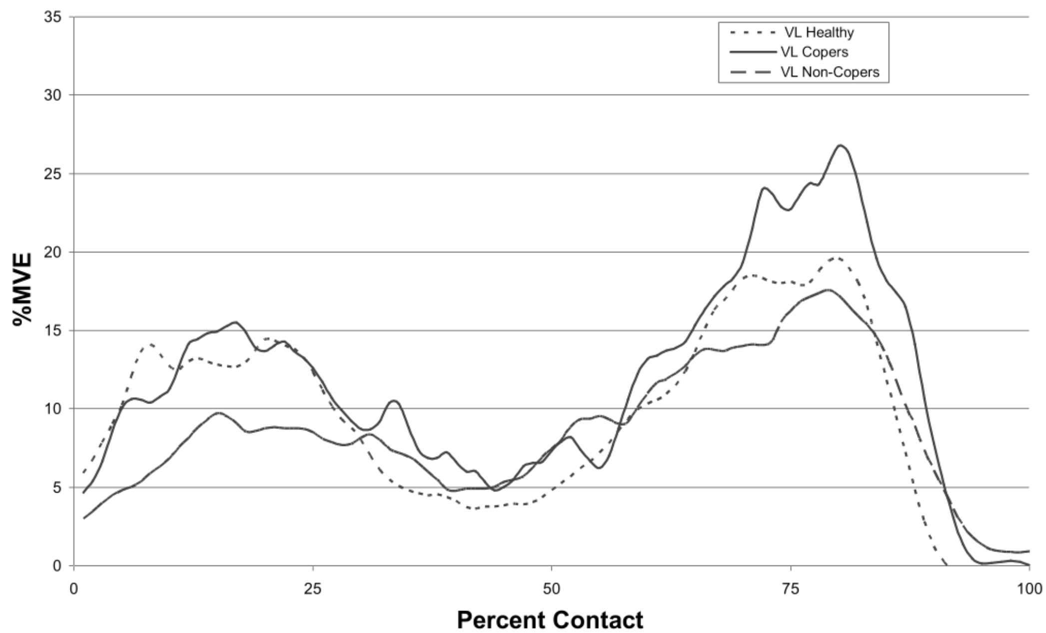

Figure 24: Vastus Lateralis Profiles during Stair Descent

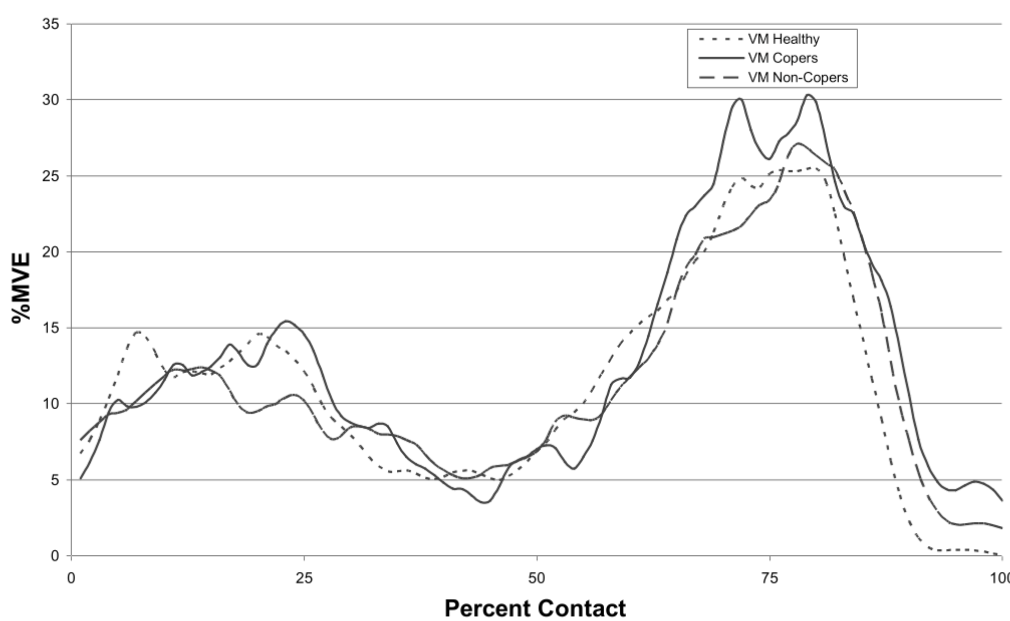

Figure 25: Vastus Medialis Profiles during Stair Descent

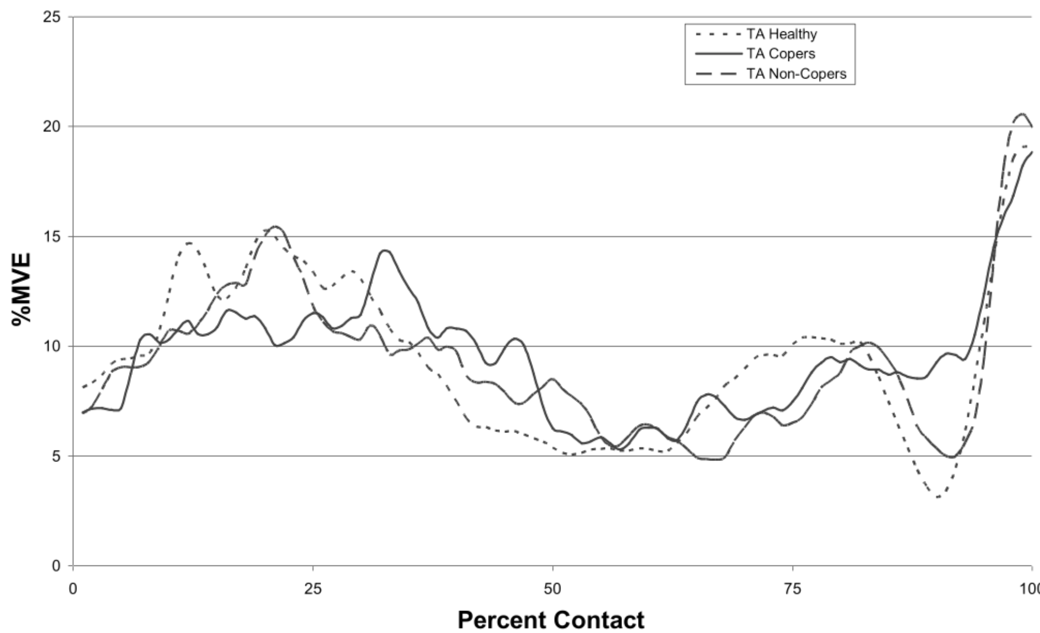

Figure 26: Tibialis Anterior Profiles during Stair Descent

4.2.1.1 Cross-Correlation

Cross-correlation analyses were performed on two sets of muscles representing the medial and lateral muscle groups acting around the knee during descent. The results of the analyses are shown in Table 11. The results refer to the correlation between the muscles at a zero time lag in an attempt to indicate the level of co-contraction on the medial and lateral sides of the knee joint. With respect to the medial side of the knee (vastus medialis/ biceps femoris), the Healthy group displayed a small positive correlation between the two muscles compared to the Non-Copers group which showed an even smaller positive correlation. Similar to ascent, the Copers group had a negative relationship between the muscles indicating that as the activity of one muscle increased the other one decreased. The differences between the groups were not found to be significantly different. With respect to the lateral side of the knee (vastus lateralis/ semitendinosus) all groups had an average negative correlation with the Healthy group having the largest correlation followed by the Copers group and then the Non-Copers. None of the differences between the groups were found to be significantly different.

Table 11: Cross Correlation Results during Stair Descent

	Copers (SD)	Non-Copers (SD)	Healthy (SD)
Vastus Medialis/ Biceps Femoris	-0.176 (0.13)	0.000 (0.43)	0.011 (0.25)
Vastus Lateralis/ Semitendinosus	-0.112 (0.28)	-0.013 (0.09)	-0.305 (0.39)

4.2.2 Kinetics

Kinetic measures were taken during descent from the start of contact on to the force plate until end of contact of the force plate. Knee and ankle moment and knee power profiles were divided into three phases of stance (early middle, and late) based upon the shape of the profiles. Based on the profile for ankle power, it was divided into four phases consisting of an a1, a2, a3, and an a4 phase. Maximum and minimum values for each group are provided in Table 12 with profiles shown in Figure 27, Figure 28, Figure 29, Figure 30, and Figure 31.

Table 12: Kinetics Results – Descent (* indicates areas of significant differences)

	Copers (SD)	Copers Range	Non-Copers (SD)	Non-Copers Range	Healthy (SD)	Healthy Range
Knee Flex-Ext Moment – Early (N-M/Kg)	2.52 (0.49)	2.22 – 3.37	2.55 (0.27)	2.10 – 2.84	2.91 (0.60)	1.75 – 3.62
Knee Flex-Ext Moment – Mid (N-M/Kg)	*1.71 (0.20)*	*1.52 – 2.05*	*2.06 (0.35)**	*1.54 – 2.53*	*1.54 (0.32)**	*0.98 – 2.02*
Knee Flex-Ext Moment – Late (N-M/Kg)	3.63 (0.60)	2.90 – 4.21	3.93 (0.43)	3.43 – 4.55	3.17 (0.48)	2.41 – 3.68
Knee Add-Abd Moment – Early (N-M/Kg)	-0.22 (0.30)	(-0.57) – 0.09	-0.22 (0.15)	(-0.38) – (-0.05)	0.09 (0.28)	(-0.57) – 0.40
Knee Add-Abd Moment – Late (N-M/Kg)	-0.37 (0.38)	(-0.72) – 0.10	-0.41 (0.29)	(-0.87) – (-0.14)	-0.27 (0.29)	(-0.67) – 0.16
Ankle Moment – Early (N-M/Kg)	0.90 (0.15)	0.68 – 1.04	0.93 (0.14)	0.75 – 1.04	0.89 (0.27)	0.54 – 1.31
Ankle Moment – Mid (N-M/Kg)	0.52 (0.14)	0.41 – 0.76	0.63 (0.10)	0.54 – 0.75	0.42 (0.14)	0.26- 0.70
Ankle Moment – Late (N-M/Kg)	0.77 (0.20)	0.50 – 1.02	0.86 (0.11)	0.70 – 0.98	0.66 (0.21)	0.34 – 0.99
Knee Power – k1 (W/Kg)	*-3.44 (2.13)*	*(-6.97) – (-1.60)*	*-1.83 (1.25)**	*(-3.09) – 0.14*	*-4.36 (1.26)**	*(-5.39) – (-1.28)*
Knee Power – k2 (W/Kg)	0.53 (0.61)	(-0.35) – 1.36	-0.33 (0.96)	(-1.26) – 1.10	1.18 (0.44)	0.47 – 1.70
Knee Power – k3 (W/Kg)	-12.60 (2.74)	(-16.9) – (-10.2)	-10.91 (5.88)	(-16.9) – (-1.64)	-10.02 (1.55)	(-12.5) – (-7.18)
Ankle Power – a1 (W/Kg)	-2.26 (0.48)	(-2.96) – (-1.66)	-2.32 (0.20)	(-2.62) – (-2.07)	-3.22 (1.12)	(-4.61) – (1.25)
Ankle Power – a2 (W/Kg)	-0.22 (0.17)	(-0.45) – (-0.07)	-0.24 (0.18)	(-0.42) – (-0.03)	-0.04 (0.12)	(-0.18) – 0.25
Ankle Power – a3 (W/Kg)	-0.77 (0.29)	(-1.23) – (-0.53)	-0.85 (0.30)	(-1.26) – (-0.42)	-0.62 (0.21)	(-0.90) – (-0.21)
Ankle Power – a4 (W/Kg)	*1.68 (0.72)*	*0.96 – 2.59*	*2.06 (0.33)**	*1.65 – 2.53*	*1.33 (0.24)**	*0.95 – 1.60*

4.2.2.1 Moments

Knee Flexion-Extension Moment

The average knee flexion-extension moment for each percent of the stance phase was calculated for each group and 3 distinct phases of the moment profile were compared between groups (early, mid, and late). The knee flexion-extension moment profiles are shown in Figure 27. The profiles illustrate that all groups maintained a net extensor moment during the entire stance phase. During early stance, the Healthy group obtained the largest net extensor moment with the Copers and Non-Copers having similar values. However, during the middle and late phases the

Non-Copers and Copers had larger net extensor moments over the Healthy group with the Non-Copers achieving the largest extensor moments. The early phase was characterized by an initial peak with the Healthy group achieving the largest peak followed by the Non-Copers and Copers. There were no statistical differences in knee moments between any of the groups during the initial phase of stance. The middle section of the stance phase was characterized by a decrease in net extensor moment with the Healthy group achieving the lowest net extensor moment followed by the Copers and Non-Copers. The Non-Copers were found to achieve a significantly greater amount of net extensor moment over the Healthy group (p-value= 0.04) but not significantly greater than the Copers. The late phase of stance is characterized by another peak net extensor moment. The Non-Copers achieved the largest net extensor moment during the late phase followed by the Copers and Healthy group. There were no significant differences in knee moments between the groups during the late phase of stance.

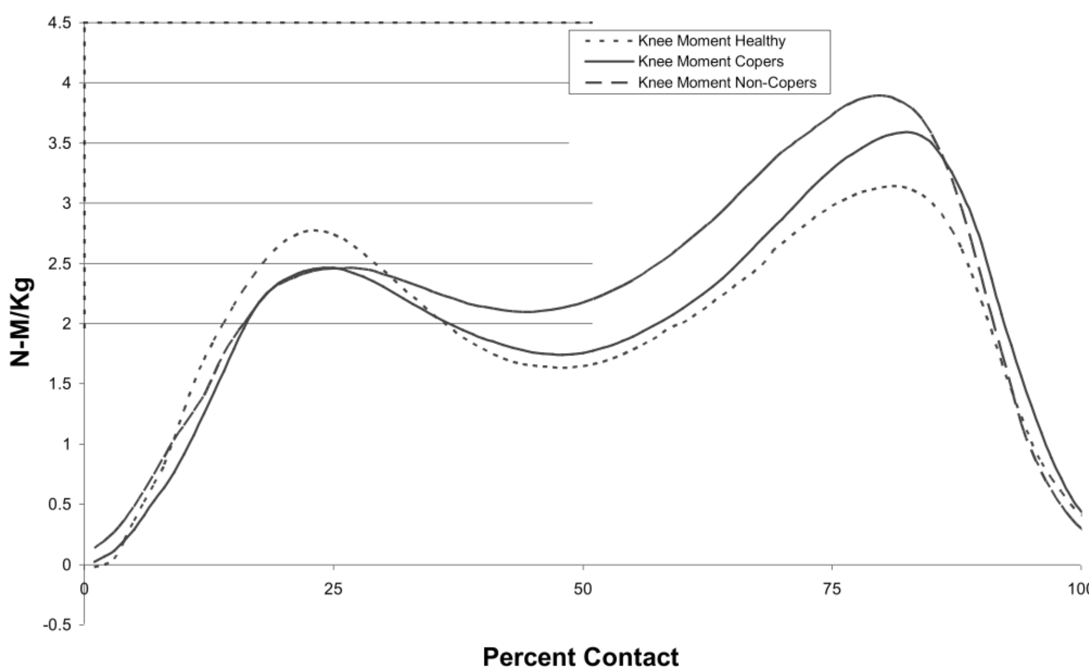

Figure 27: Knee Flexion-Extension Moment Profiles during Stair Descent (positive values indicate extensor moments and negative values indicate flexor moments)

Knee Abduction-Adduction Moment

The average knee adduction-abduction moment for each percent of the stance phase was calculated for each group and 2 distinct phases of the moment profile were compared between groups (early and late). The knee adduction-abduction moment profiles are shown in Figure 28. The ACL-D groups maintained an abduction moment on the knee throughout the entire stance phase, while the Healthy group had an adduction moment through the first 30% of stance and transferred into an abduction moment for the duration of the stance phase. The adduction-abduction moment profile is characterized by 2 peaks with the first peak being smaller than the second peak. At the onset of contact onto the force plate the Copers and Non-Copers groups had approximately the same level of abduction moment, while the Healthy group had a small adduction moment. Throughout the majority of the remaining portion of the stance phase, the Non-Copers group maintained a larger abduction moment followed by the Copers group and then the Healthy group. At the point at which the stance phase ended, the Copers and Non-Copers groups had very similar abduction moments while the Healthy group had a slightly lower abduction moment. The maximum adduction-abduction moment during the first and second peaks in the moment profile were found for each participant and compared across groups. The first peak took place between 15 – 40% of the stance phase. As previously mentioned, the Non-Copers group had the largest lateral moment with a value of 0.222 N-M/Kg, followed by the Copers group with a value of 0.219 N-M/Kg, and then by the Healthy group with a value of 0.088 N-M/Kg. The differences between the groups during the early peak were not found to be statistically significant. The second peak took place between 65 – 95% of the stance phase with the Non-Copers group achieving the largest lateral moment with a value of 0.414 N-M/Kg, then by the Copers group with a value of 0.366 N-M/Kg, and then by the Healthy group with a value of 0.269 N-M/Kg. The differences between the groups during the late peak were not found to statistically significant.

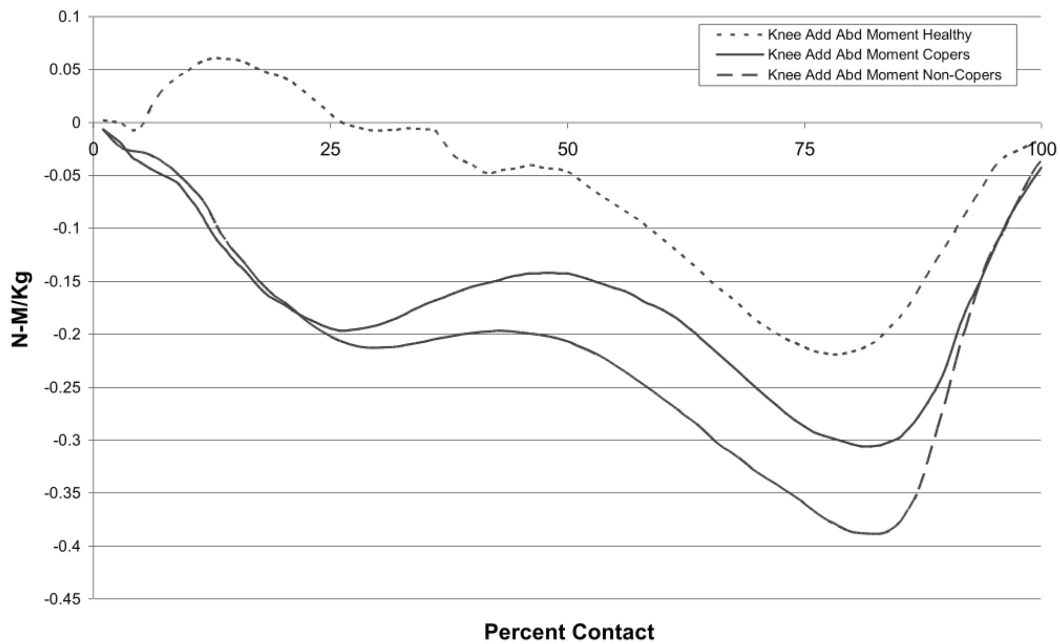

Figure 28: Adduction and Abduction Knee Moments during Stair Descent (positive values indicate abduction moments and negative values indicate adduction moments)

Ankle Flexion-Extension Moments

The average ankle moment for each percent of the stance phase was calculated for each group and 3 distinct phases of the moment profile was compared between groups (early, mid, and late). The ankle moment profiles are shown in Figure 29. The profiles illustrate all the groups achieved consistent net plantarflexor moments during all the phases of stance. The Non-Copers group achieved larger peaks during the early and late portions of stance and larger values during the mid portion of stance where the net plantarflexor moments decreased. During the early phase the Non-Copers group achieved the largest plantarflexor moment, followed by the Copers group, and then by the Healthy group. The difference between the groups during the early phase was not found to be statistically different. The middle phase represented a decreased net plantarflexor moment with the Healthy group achieving the smallest plantarflexor moment followed by the Copers group and then by the Non-Copers group. The differences between groups during mid stance were not found to be statistically different. During the late phase all groups increased their net plantarflexor moments with the Non-Copers group achieving the highest value followed by the Copers group and finally then Healthy group. The differences between the groups were not found to be statistically different.

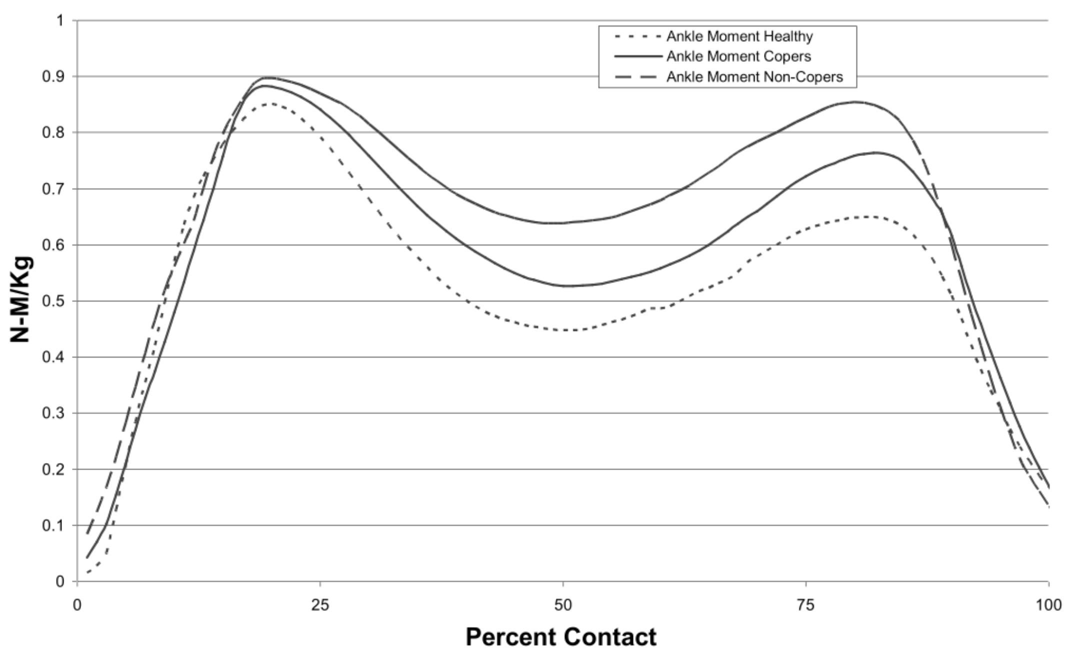

Figure 29: Ankle Moment Profiles during Stair Descent (positive values refer to plantarflexor moments and negative values indicate dorsiflexor moments)

4.2.2.2 Powers

Knee Power

The average knee power for each percent of the stance phase was calculated for each group and 3 distinct phases of the power profile were compared between groups (k1, k2, and k3). The knee power profiles are shown in Figure 30. The profiles illustrate a period of eccentric activity by the knee extensors for the first 25% of stance followed by another brief period of concentric activity for the Copers and Healthy group between 25 – 50% of stance, and then the last half of the stance phase is a long period of increasing eccentric activity to approximately 90% of stance and then a sharp decrease of eccentric activity until toe off. The Non-Copers group remain in eccentric contraction for the entire duration of stance. The Non-Copers follow the same knee power profile as the other groups but they never obtain a period of concentric contraction between 25 – 50% of the stance phase. During k1, the Healthy group achieved the largest amount of eccentric activity, followed by the Copers group, and then the Non-Copers group. During k1, the Healthy group was found to achieve a significantly greater peak amount of eccentric activity over the Non-Copers (p-value=0.042040). The k2 represented a decrease in eccentric activity and shift to concentric activity for the Copers and Healthy group. The Healthy group achieved the largest amount of concentric activity, followed by the Copers group, and then by the Non-Copers with an eccentric activation value of -0.307 W/Kg. The difference between the groups during the middle phase was not found to be statistically different. The k3, from 50% to 100% stance, had a change of concentric activity to eccentric activity for the Copers and Healthy groups and an increase of eccentric activity for the Non-Copers. Maximum values during this phase occurred at approximately 90% of stance with the Copers group achieving the largest mean peak eccentric activity, followed by the Non-Copers group, and then followed by the Healthy group. The differences between the groups in the late phase were not found to be statistically different.

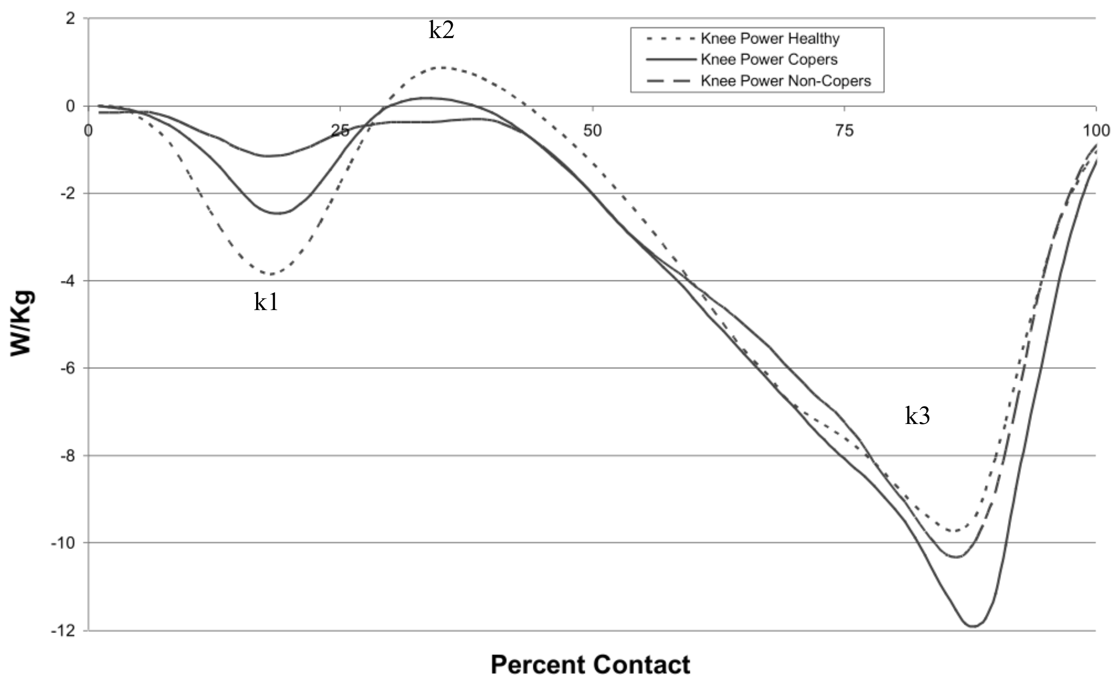

Figure 30: Knee Power Profiles during Stair Descent (positive values indicate power generation and negative values indicate power absorption)

Ankle Power

The average ankle power for each percent of the stance phase was calculated for each group and 4 distinct phases of the power profile were compared between groups (a1, a2 and a3, and a4). The ankle power profiles are shown in Figure 31. The profiles illustrate a large period of eccentric activity by the plantarflexors for the first 80% of stance that consisted of two peaks and a single minimum followed by a brief period of concentric activity, by the plantarflexors, during the last 20% of stance. The a1 phase consisted of the onset of stepping onto the force plate to approximately 25% of the stance phase and consisted of a large peak in eccentric activity. The Healthy group achieved the largest peak eccentric activity, followed by the Non-Copers and Copers. The a2 phase took place between 25% and 50% of the stance phase with a decrease amount of eccentric activity with the Healthy group achieving the lowest amount of eccentric activity followed by the Copers and Non-Copers. The a3 phase was found during 50% and 80% of the stance phase. This period featured an increase in eccentric activity with the Non-Copers group having the largest amount of eccentric activity, followed by the Copers and the Healthy group. The a4 of stance featured a brief period of concentric activity with the Non-Copers group having the largest amount of concentric activity followed by the Copers and the Healthy group. No statistical differences were found between any of the groups during the first three phases of the stance phase. During the a4 the Non-Copers group had a significantly larger amount of peak concentric activity compared to the Healthy group (p-value=0.04).

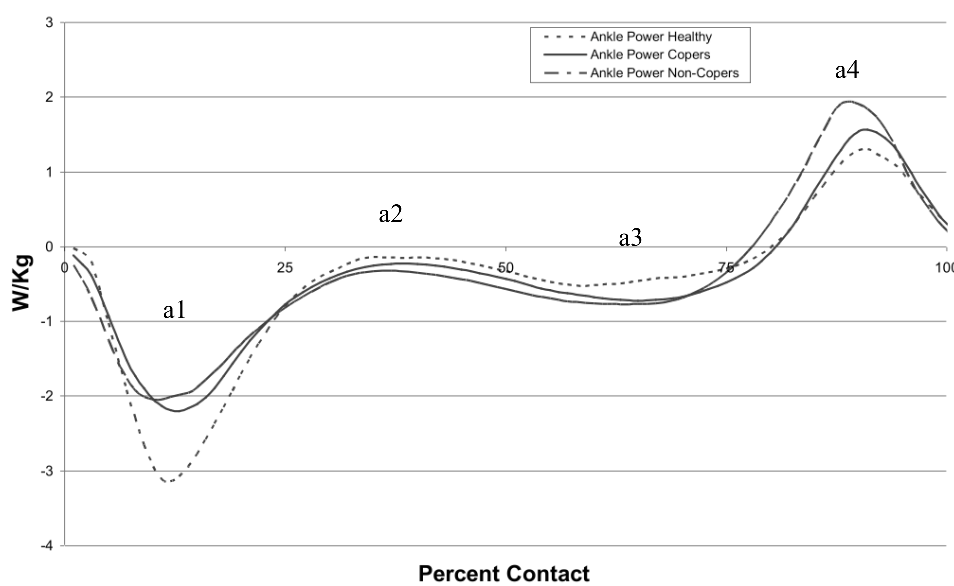

Figure 31: Ankle Power Profiles during Stair Descent (positive values indicate power generation and negative values indicate power absorption)

4.2.3 Kinematics

Knee angles were measured for each percentage of stance and the profiles are shown in Figure 32. All groups contacted the force plate with their knee in an extended position and during the progression to the next step their knee became increasingly flexed. Five instances were selected for comparison across groups (onset, 25% stance, 50% stance, 75% stance, and offset) and the results are shown in Table 13. In each instance, the Non-Copers group was approximately 1- 7 degrees more flexed than the Copers group and anywhere from 4 – 7 degrees more flexed than the Healthy group until toe-off where the Non-Copers were approximately 16 degrees more flexed at the knee compared to the Healthy group. The Copers group was more flexed than the Healthy group for the majority of the stance phase and slightly more extended during the early part of stance. Similar to stair ascent, the biggest differences were observed between the groups at toe-off where the Non-Copers knees were flexed on average 9 degrees more than the Copers group and 16 degrees more than the Healthy group. Even though differences were observed between the groups, none of them were found to be statistically significant at any of the key instances selected.

Table 13: Knee Angles at Specific Times during Stair Descent

	Copers (SD)	Copers Range	Non-Copers (SD)	Non-Copers Range	Healthy (SD)	Healthy Range
Onset – (degrees)	16.5 (5.6)	16.5 – 26.7	22.3 (9.2)	13.0 - 36.7	18.0 (5.3)	11.4 – 24.3
25% Stance – (degrees)	26.6 (11.0)	25.6 – 49.8	32.1 (7.6)	23.7 – 42.4	28.9 (5.2)	21.9 – 40.7
50% Stance – (degrees)	29.5 (6.3)	30.4 – 38.8	35.0 (9.4)	21.9 – 51.2	28.2 (5.3)	24.1 – 41.2
75% Stance – (degrees)	56.0 (4.8)	50.3 – 68.7	61.3 (5.8)	40.5 – 66.4	55.1 (10.4)	37.6 – 69.1
Offset – (degrees)	95.2 (3.9)	89.5 – 109.0	103.0 (4.0)	63.2 – 103.8	85.7 (17.5)	60.6 – 103.8

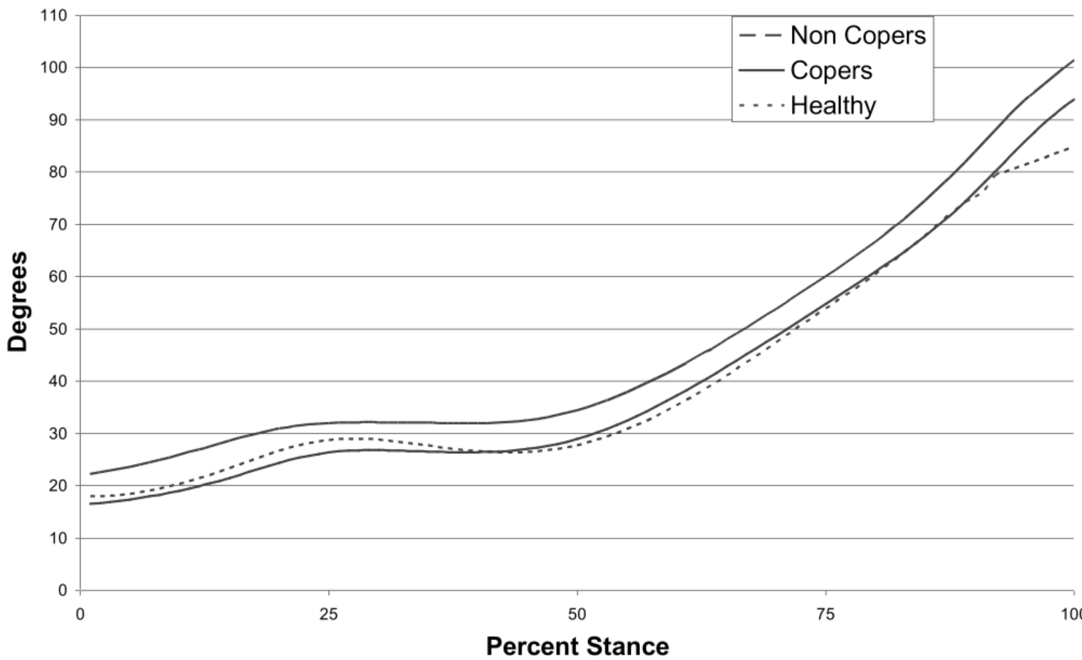

Figure 32: Knee Angle Profiles during Stair Descent (full at 0 degrees)

5. Discussion

The purpose of this study was to identify the control strategies used by ACL-deficient Copers and Non-Copers during stair climbing tasks. These control strategies were compared to a healthy population in an attempt to identify potential control strategies that may accelerate the onset of osteoarthritis following rupture of an ACL. It was originally hypothesized that the dynamic control strategies used by ACL-D Copers and Non-Copers would differ from those of the healthy group. In fact, out of the 52 variables measured, only 6 variables were found to be significantly different (3 ascent and 3 descent). However, a number of trends and common differences were found between the ACL-D groups and the healthy group that may lend insight into how altered joint mechanics may lead to a degenerative process.

5.1 Ascent

At the time of this report there were no studies that had looked at the differences between ACL-D Copers and Non-Copers during stair climbing so a direct comparison of previous results with this study cannot be made. Studies with other tasks have been conducted and can be compared to the results found in this study. During both stair ascent and descent, there were no significant differences between the groups at any of the specific instances that were chosen for comparison, however trends were observed between groups. The Non-Copers group had greater knee flexion than the Coper and healthy groups throughout both stair ascent and descent. During ascent, this may be a stabilizing mechanism used by the Non-Copers since the hamstring muscle is placed at a greater mechanical advantage at larger flexion angles and is able to compensate better for a deficient ACL at this joint angle (Robert et al. 1999). When examining the EMG of the hamstring muscles it shows that the Non-Copers group had elevated Biceps Femoris and Semitendinosus activity throughout the duration of stance compared to the other groups. Both the Non-Copers and the Copers groups had a more variable EMG signal compared to the Healthy group. This may be attributable to the ACL-D groups making constant adjustments, based on feedback, during stance to accommodate for the lack of an ACL. However, when examining the individual graphs it appears this is not the case and may be due to the fact that there were 10 Healthy participants compared to 5 people in each of the ACL-D groups.

There was no reduction in quadriceps activity by either the Copers group or the Non-Copers group during the stair ascent suggesting that hamstring facilitation by the Non-Copers group was the key control strategy by the Non-Copers group for attempting to maintain knee joint stability. During the late stages of stance (approximately 80% of stance) the Healthy and Non-Copers groups show an increase in quadriceps activity while the Copers group essentially have turned off the quadriceps relying solely on the gastrocnemius muscles for ascent. At this instance the knee is approximately at 20° of flexion where activity of the quadriceps muscle produces more of an anterior shear force on the knee joint and the underlying cartilage. While this might not impact the Healthy group as they have an intact ACL that can reduce this anterior force, the Non-Copers group is more susceptible to this anterior tibial translation. This increase in quadriceps activity in the Non-Copers group is also coupled with an extensor moment at the knee while the other groups maintain a flexor moment.

Previous research has shown that ACL-D groups reduce the net extensor moment at the knee as coping strategy for the lack of an ACL. It was hypothesized that the Non-Copers would reduce their extensor moment about the knee in an attempt to reduce any unwanted anterior tibial translation. Throughout the majority of the stance phase the Copers and the Non-Copers group showed an overall decrease in flexion-extension moment at the knee, with the Copers group having a greater reduction in extensor moment compared to the Healthy group, and the Non-Copers group having a greater reduction in flexor moment. McFadyen and Winter (1988), stated at approximately 25 – 30% of stance there is an increase in concurrent activity of the extensor muscles at the knee and ankle as the opposite foot begins the swing phase and the stance limb takes all of the weight. At this instance the Copers group and the Healthy group have almost identical knee flexor moments, while the Non-Copers have a reduced flexor moment. This suggests that at the key instance where the stance limb takes all of the weight the Copers group is able to maintain knee moments of the same magnitude as the Healthy group. Research done by Protopapadaki et al. (2006) showed similar knee moment flexion-extension profiles during stair ascent to the results in this study. In both studies subjects contacted the force plate with a net knee flexion moment and then proceeded with a brief net knee extensor moment that took place during the initial 30% of the stance phase. After this, healthy participants in both studies shifted to a large net knee flexion moment. However, in our study this lasted for the duration of the contact phase of the force plate. Protopapadaki et al. (2006) found that participants briefly shifted to a net knee extensor moment between 70 – 80% of the stance phase before returning to a net knee flexion moment. We did not observe this late shift to a net knee extensor moment within our healthy group, however we did observe a brief shift to a net knee extensor moment among our Non-Copers group that took place between 90 – 98% of the stance phase. The knee moment values calculated during ascent in both our study and the study conducted by Protopapadaki et al. (2006) were fairly similar with peak knee extensor moment values of approximately 0.45 N·M/Kg in the previous study compared approximately 0.3 N·M/Kg in our study. Peak knee flexor values were also similar with values in the previous study around 0.6 N·M/Kg compared to 0.8 N·M/Kg in the current study. The differences between the current study and the Protopapadaki et al. (2006) could be due to the differences in stair dimensions. The stairs in the this study were slightly higher (1.5 cm higher) and slightly wider (1.5 cm wider) which may account for the some of the differences. During the first 15% of stance the Copers and Healthy groups do not show a smooth transition from a flexor moment to an extensor moment. This was due to 2 participants in each group who had changed from flexor moments to extensor moments quickly. Also, a single Coper maintained a net flexor moment throughout the entire stance phase. Not only did the Copers and Non-Copers groups have an overall decrease in flexion-extension moments but they also had an overall decrease in knee abduction moments for the majority of the stance phase compared to the Healthy group.

Stair ascent is typically characterized by the concentric activation of many of the extensor muscles of the leg in order to pull and push the body up and over the step (McFadyen and Winter, 1988). Theoretically this should result in the generation of power throughout the step cycle. However, in our study the Healthy and Non-Copers groups had a brief period of energy absorption by the knee extensors during the first 30% of stance while the Copers generated energy through the first 85% of stance. Two out of the five Copers participants showed similar profiles to the Non-Copers and Healthy groups, however the remaining three Copers participants had large power generation throughout the first 85% of stance. After the first 30% of stance, both the Non-Copers and Healthy groups produced power at the knee to ascend the step with all 3 groups displaying similar profiles. Copers also showed lower extensor moments during the early part of stance. This indicates a strategy used by the Copers to which during the early part of stance they can generate concentric power at the knee (to pull and push the body to the next step) and still maintain a lower extensor moment at the knee.

Therefore, to accomplish stair ascent Non-Copers maintained a greater amount of knee flexion with an overall reduction in knee flexion-extension moment. They also had increased hamstring muscle activity, which may lead to increase in joint contact forces, due to the increase in muscle forces. These potential increases in joint forces may be placed on articular cartilage that is not used to be loaded to that extent due to the increases in knee flexion angle throughout the duration of the stance phase. The Non-Copers may be successful at accomplishing the stair ascent task with greater responsibility at the hip.

DeVita and Hortobagyi (2000) looked at differences in lower limb mechanics between an elderly group and healthy individuals and showed that elderly individuals increase extensor output at the hip as a compensation for lower extensor output at the knee (DeVita and Hortobagyi, 2000). There is also evidence that shows a redistribution of joint moments among ACL injured individuals. DeVita et al. (1998) showed that the support moment between ACL injured individuals and healthy individuals were the same, even though there were less power and moments generated at the knee. Power and net joint moments at the hip were increased to compensate for the knee. Since we did not collect data at the hip we could not test this assumption.

However, there was a lack of statistical evidence during ascent that would suggest that Non-Copers and Copers actually ascend stairs differently. Costigan et al. (2002) measured the patello-femoral contact force during stair ascent and found it to be directed backward, thus relieving strain in the anterior direction, thus allowing ACL-D subjects to perform the task normally. This assumption may be statistically true but this study has identified trends in control strategies used by the Non-Copers during ascent which may place them at an increased risk of articular cartilage breakdown and the potential for onset of osteoarthritis.

5.2 Descent

During stair descent the Non-Copers group maintained greater knee flexion angles over the other groups throughout the course of stance. The Copers group had very similar knee angles to the Healthy group throughout stance. Even though there were no significant differences between the groups, in terms of knee angle, it does identify that the Non-Copers tend to descend stairs in a more flexed knee pattern. During stair descent the quadriceps muscle eccentrically contracts to control the lowering of the body and leg. Non-Copers who would avoid quadriceps activity, as previous research suggests, would lower the body to greater levels thus increasing the knee flexion angle.

Since descent is more of power absorbing task and muscles about the knee are typically not used to generate large bursts in power smaller differences between the groups were observed with respect to muscle activation. There were no significant differences between any of the groups for any of the variables we measured. However, we did notice that the Non-Copers group did have more hamstring activity throughout the duration of the stance phase signified by their AEMG levels. Throughout the majority of the stance phase Copers and Non-Copers had increased levels of biceps femoris and semitendinosus activity as a potential means of controlling displacement of the tibia. This was more evident with the Non-Copers who had a higher level of activation throughout the entire step cycle, except for the last 10% of stance. The activation patterns of the semitendinosus and biceps femoris show that for Non-Copers tibial translation is controlled by both muscles at different times of the step cycle. The biceps femoris works mainly at the beginning and end of the cycle while the semitendinosus is more active during the middle part of stance.

This increase in hamstring muscle activity by the Non-Copers, without any significant decrease in quadriceps activity, places greater forces on the knee at different knee angle, when compared to the healthy group. Potentially the Non-Copers will be loading different sites of articular cartilage at greater levels and could promote the initiation of cartilage breakdown and the initiation of osteoarthritis.

The Copers group quadriceps muscle activity was more variable due to the differences between the subjects rather than constant variations in EMG within each subject. The Non-Copers groups show the same variation in the hamstring muscles between subjects rather greater variation in signal within each subject.

As opposed to stair ascent, in stair descent we did not see the same overall reduction in knee flexion-extension moment as we saw during stair ascent. The ACL-D groups had almost the same amount of extensor moments for the first 25% of stance where the Healthy groups had slightly higher amounts of extensor moments. Between 25 - 75% of stance, the Non-Copers group had the higher levels of extensor moment with the Healthy group having the lowest levels. At approximately 50% of stance the Non-Copers had significantly greater amounts of knee extensor moment compared to the Healthy group. The increase in knee moment in the Non-Copers provides further evidence that the Non-Copers are placing greater amounts of force on the knee joint cartilage. The values for knee moments in the flexion-extension plane in this study were quite larger compared to the ones by Protopapadaki et al. (2006). In our study, peak knee extensor moments for the healthy group were around 3.1 N·M/kg while in the Protopapadaki et al. (2006) study, the average peak knee extensor moment values were around 0.4 N·M/Kg. That is a large difference in knee moments compared to the values calculated in our study but are more similar to the knee moments calculated in the study by McFadyen and Winter (1988) where they had peak internal extensor moments around 1.5 – 2.0 N·m/Kg. Again differences between the studies could be due to the differences in the dimensions of the staircase.

In the abduction-adduction plane the knee moment profiles between the groups were vastly different . The ACL-D groups consistently had an adduction moment at the knee throughout the entire stance phase. The healthy group had an adduction moment through the last 70% of stance but during the first 30% of stance they mostly had an abduction moment at the knee. This difference was not found to be significant based on peak values, but it does signify that the ACL-D groups may have a tendency of loading the lateral side of the knee more than the healthy groups during descent. This increase in lateral loading will also play a factor in the early development of OA, since it is interrupting the normal physiological loading causing increased loading of cartilage that is typically underloaded which may elicit degenerating changes to the articular cartilage (Wu et al. 2000).

Stair descent appears to pose a larger risk to articular cartilage degeneration in ACL-D groups, particularly the Non-Copers. Not only do the Non-Copers maintain a more flexed knee angle, but they also increase the moments about the knee in both the flexion-extension plane and the abduction-adduction plane. These increases in moments about the knee and the increase in hamstring muscle activity will most likely place greater contact forces at the knee and overload specific areas of articular cartilage.

5.3 Limitations

No study goes without limitations. One of the key limitations of this study was the small sample size of our ACL-D groups. A total of only 5 participants for each of the ACL-D group were found. A larger sample size may have identified significant differences between the groups where in this case were not revealed. Another limitation of this study was the inability to perform functional testing of all of the ACL-D participants as means of classifying them as either a Coper or Non-Coper. Hopping tests were found to be too demanding for some of the ACL-D subjects and thus they were removed from the protocol in an attempt to decrease the risk of injury to the participants. Instead, the experimenters relied on subjective assessments as a means of classifying individuals into either the Coper or Non-Coper groups.

Another limitation to this study was that kinetic and kinematic data of the hip were not collected due to the fact that this study was only concerned about the strategies used at the knee. The ankle was necessary for these calculations so therefore it was included in our results and discussion. If this study was focused on the lower limb differences between groups then the hip would need to be included. With that said the hip may have played an integral part in the compensation mechanisms used by the ACL-D groups to successfully complete the stair climbing task. Few differences were observed at the ankle so it can be assumed that differences may have been observed at the hip to compensate for changes at the knee joint. In future studies, kinetic and kinematic data at the hip should be collected.

Another limitation is that most of the variables selected for comparison were single values chosen from the group based on key instances during stair ascent and descent. For example, knee kinematic values were chosen at incriments of 25% of the stance phase. Therefore, it cannot be assumed that differences between the groups do not exist between the points selected, and significant differences may have been missed. Another method of examining the curves more closely would be to take the area under the curves between selected points. This would provide a gross estimate of the pattern of the variable between the groups.

Finally, we did not take into account leg dominance in the ACL-D groups as we could not control the limb that was injured. Ounpuu and Winter (1989), identified EMG asymmetries in the limbs. This may be due to the differences in muscle volume and muscle activity of the dominant limb to perform the same task as the non-dominant limb. In the case of ACL-D we can assume that the muscle actiity seen between the limbs goes beyond dominance and serves as a compensatory mechanism to achieve task goals without the presence of an ACL. Hamill et al. (1984) found no statistical difference between the dominant and non-dominant limb when ground reaction force parameters were measured. Even though the gait literature suggests that there is asymmetry between the limbs depending on the function of each limb during the gait cycle (either lead limb or trail limb), this is not suggested during the stair climbing literature where the task of stair climbing is very different compared to gait. In gait, one limb is used for propulsion while the other is used for stability. In stair climbing it is assumed that both limbs are equally responsible for asceding or descending the stairs independently (Hamill et al. 1984).

6. Conclusions

The main goal of this study was to identify any differences in the control strategies used by ACL-D Copers and Non-Copers during stair climbing. The stair climbing task was chosen as it is a common task that is encountered on a daily basis and is a more demanding task than walking.

We anticipated the Copers group would ascend and descend the stairs with their knee angle more flexed than the Non-Copers in an attempt to place the hamstring muscle at a greater mechanical advantage to resist the anterior tibial translation caused by the quadriceps. In this study however, we found that the Non-Copers were the group that had their knee in greater flexion. The Copers group tended to use the plantarflexors of the ankle to ascend the stairs and less of the knee extensor muscles. This may reduce the need to provide of the hamstring to obtain a mechanical advantage to resist anterior tibial translation.

We also anticipated that the Copers group would co-contract muscles to a greater extent during stair climbing. However, in this study we did not find any differences between the groups with respect to co-contraction. The Non-Copers group had slightly more co-contraction than the Copers group but the differences were not significant based on our cross-correlation analysis.

We measured a total of 52 variables during both ascent and descent and found a total of only 6 significant differences between the groups. During ascent 2 variables were found to be statistically different. During the late phase of stance the Non-Copers exhibited a net internal extensor moment at the knee, produced by the quadriceps muscles, compared to the healthy group which showed a net internal flexor moment. This significant difference shows that the Non-Copers may put their knee at a vulnerable position near the late stages of stance due to the internal extensor moment at the knee. The healthy group also had significantly larger peak activation of the Medial Gastrocnemius and the Vastus Medialis muscles during ascent over the Copers group. Even though, the Copers group did not activate the Medial Gastrocnemius to the same level of the healthy group they did have a larger AEMG signifying that the Medial Gastrocnemius muscle was more active during the entire duration of stance as opposed to one exact moment. The Copers group also did not activate a key power generator during stair ascent, in the form of the Vastus Medialis, as a possible attempt to limit the potential anterior tibial translation at the knee and possibly decrease the likelihood of instability.

During descent there were also 3 variables that were found to significantly different among the groups. At the early stages of stance the healthy group was able to absorb significantly more energy at the knee then the Non-Copers and during the late stages of stance the Non-Copers were able to absorb more energy at the ankle over the healthy group. This suggests that the Non-Coper group may not be able to absorb energy at the knee as efficiently at the knee, due to quadriceps activation, and therefore must transfer the absorption of energy to the muscles surrounding the ankle during the late stages of stance. During the middle stages of stance the Non-Copers also exhibited a significantly greater internal extensor moment at the knee compared to the healthy group signalling that the Non-Copers produces increased amount of quadriceps activity during the descent phase.

This study tried to lend insight to the development of osteoarthritis following joint injury. Even though we did not get a significant large sample of injured subjects we did find some significant differences between the groups indicating that the loading at the knee is different between the groups and may have an affect on the future development of osteoarthritis due to altered joint loading patterns.

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Appendix 1: Questionnaire

Knee Outcome Survey

Activities of Daily Living Scale
Instructions: The following questionnaire is designed to determine the symptoms and limitations that you experience because of your knee while you perform your usual *daily activities*. Please answer each question by *checking the statement that best describes you over the last 1 to 2 days.* For a given question, more than one of the statements may describe you, but please mark ONLY the statement that best describes you during your usual daily activities.
Symptoms
1. To what degree does pain in your knee affect your daily activity level?
_5_ I never have pain in my knee.
_4_ I have pain in my knee, but it does not affect my daily activity.
_3_ Pain affects my activity slightly.
_2_ Pain affects my activity moderately.
_1_ Pain affects my activity severely.
_0_ Pain in my knee prevents me from performing all daily activities.

2. To what degree does grinding or grating of your knee affect your daily activity level?
_5_ I never have grinding or grating in my knee.
_4_ I have grinding or grating in my knee, but it does not affect my daily activity.
_3_ Grinding or grating affects my activity slightly.
_2_ Grinding or grating affects my activity moderately.
_1_ Grinding or grating affects my activity severely.
_0_ Grinding or grating in my knee prevents me from performing all daily activities.

3. To what degree does stiffness in your knee affect your daily activity level?
_5_ I never have stiffness in my knee.
_4_ I have stiffness in my knee, but it does not affect my daily activity.
_3_ Stiffness affects my activity slightly.
_2_ Stiffness affects my activity moderately.
_1_ Stiffness affects my activity severely.
_0_ Stiffness in my knee prevents me from performing all daily activities.

4. To what degree does swelling in your knee affect your daily activity level?
_5_ I never have swelling in my knee.
_4_ I have swelling in my knee, but it does not affect my daily activity.

_3_ Swelling affects my activity slightly.
_2_ Swelling affects my activity moderately.
_1_ Swelling affects my activity severely.
_0_ Swelling in my knee prevents me from performing all daily activities.

5. To what degree does slipping of your knee affect your daily activity level?
_5_ I never have slipping of my knee.
_4_ I have slipping of my knee, but it does not affect my daily activity.
_3_ Slipping affects my activity slightly.
_2_ Slipping affects my activity moderately.
_1_ Slipping affects my activity severely.
_0_ Slipping of my knee prevents me from performing all daily activities.

6. To what degree does buckling of your knee affect your daily activity level?
_5_ I never have buckling of my knee.
_4_ I have buckling of my knee, but it does not affect my daily activity level.
_3_ Buckling affects my activity slightly.
_2_ Buckling affects my activity moderately.
_1_ Buckling affects my activity severely.
_0_ Buckling of my knee prevents me from performing all daily activities.

7. To what degree does weakness or lack of strength of your leg affect your daily activity level?
_5_ My leg never feels weak.
_4_ My leg feels weak, but it does not affect my daily activity.
_3_ Weakness affects my activity slightly.
_2_ Weakness affects my activity moderately.
_1_ Weakness affects my activity severely.
_0_ Weakness of my leg prevents me from performing all daily activities.

Functional Disability with Activities of Daily Living
8. How does your knee affect your ability to walk?
_5_ My knee does not affect my ability to walk.
_4_ I have pain in my knee when walking, but it does not affect my ability to walk.
_3_ My knee prevents me from walking more than 1 mile.
_2_ My knee prevents me from walking more than 1/2 mile.
_1_ My knee prevents me from walking more than 1 block.
_0_ My knee prevents me from walking.

9. Because of your knee, do you walk with crutches or

a cane?
_3_ I can walk without crutches or a cane.
_2_ My knee causes me to walk with 1 crutch or a cane.
_1_ My knee causes me to walk with 2 crutches.
_0_ Because of my knee, I cannot walk even with crutches.

10. Does your knee cause you to limp when you walk?
_2_ I can walk without a limp.
_1_ Sometimes my knee causes me to walk with a limp.
_0_ Because of my knee, I cannot walk without a limp.

11. How does your knee affect your ability to go up stairs?
_5_ My knee does not affect my ability to go up stairs.
_4_ I have pain in my knee when going up stairs, but it does not limit my ability to go up stairs.
_3_ I am able to go up stairs normally, but I need to rely on use of a railing.
_2_ I am able to go up stairs one step at a time with use of a railing.
_1_ I have to use crutches or a cane to go up stairs.
_0_ I cannot go up stairs.

12. How does your knee affect your ability to go down stairs?
_5_ My knee does not affect my ability to go down stairs.
_4_ I have pain in my knee when going down stairs, but it does not limit my ability to go down stairs.
_3_ I am able to go down stairs normally, but I need to rely on use of a railing.
_2_ I am able to go down stairs one step at time with use of a railing.
_1_ I have to use crutches or a cane to go down stairs.
_0_ I cannot go down stairs.

13. How does your knee affect your ability to stand?
_5_ My knee does not affect my ability to stand. I can stand for unlimited amounts of time.
_4_ I have pain in my knee when standing, but it does not limit my ability to stand.
_3_ Because of my knee I cannot stand for more than 1 hour.
_2_ Because of my knee I cannot stand for more than 1/2 hour.
_1_ Because of my knee I cannot stand for more than 10 minutes.

_0_ I cannot stand because of my knee.

14. How does your knee affect your ability to kneel on the front of your knee?
_5_ My knee does not affect my ability to kneel on the front of my knee. I can kneel for unlimited amounts of time.
_4_ I have pain when kneeling on the front of my knee, but it does not limit my ability to kneel.
_3_ I cannot kneel on the front of my knee for more than 1 hour.
_2_ I cannot kneel on the front of my knee for more than 1/2 hour.
_1_ I cannot kneel on the front of my knee for more than 10 minutes.
_0_ I cannot kneel on the front of my knee.

15. How does your knee affect your ability to squat?
_5_ My knee does not affect my ability to squat. I can squat all the way down.
_4_ I have pain when squatting, but I can still squat all the way down.
_3_ I cannot squat more than 3/4 of the way down.
_2_ I cannot squat more than 1/2 of the way down.
_1_ I cannot squat more than 1/4 of the way down.

_0_ I cannot squat at all.

16. How does your knee affect your ability to sit with your knee bent?
_5_ My knee does not affect my ability to sit with my knee bent. I can sit for unlimited amounts of time.
_4_ I have pain when sitting with my knee bent, but it does not limit my ability to sit.
_3_ I cannot sit with my knee bent for more than 1 hour.
_2_ I cannot sit with my knee bent for more than 1/2 hour.

Appendix 2: Ascent Graphs

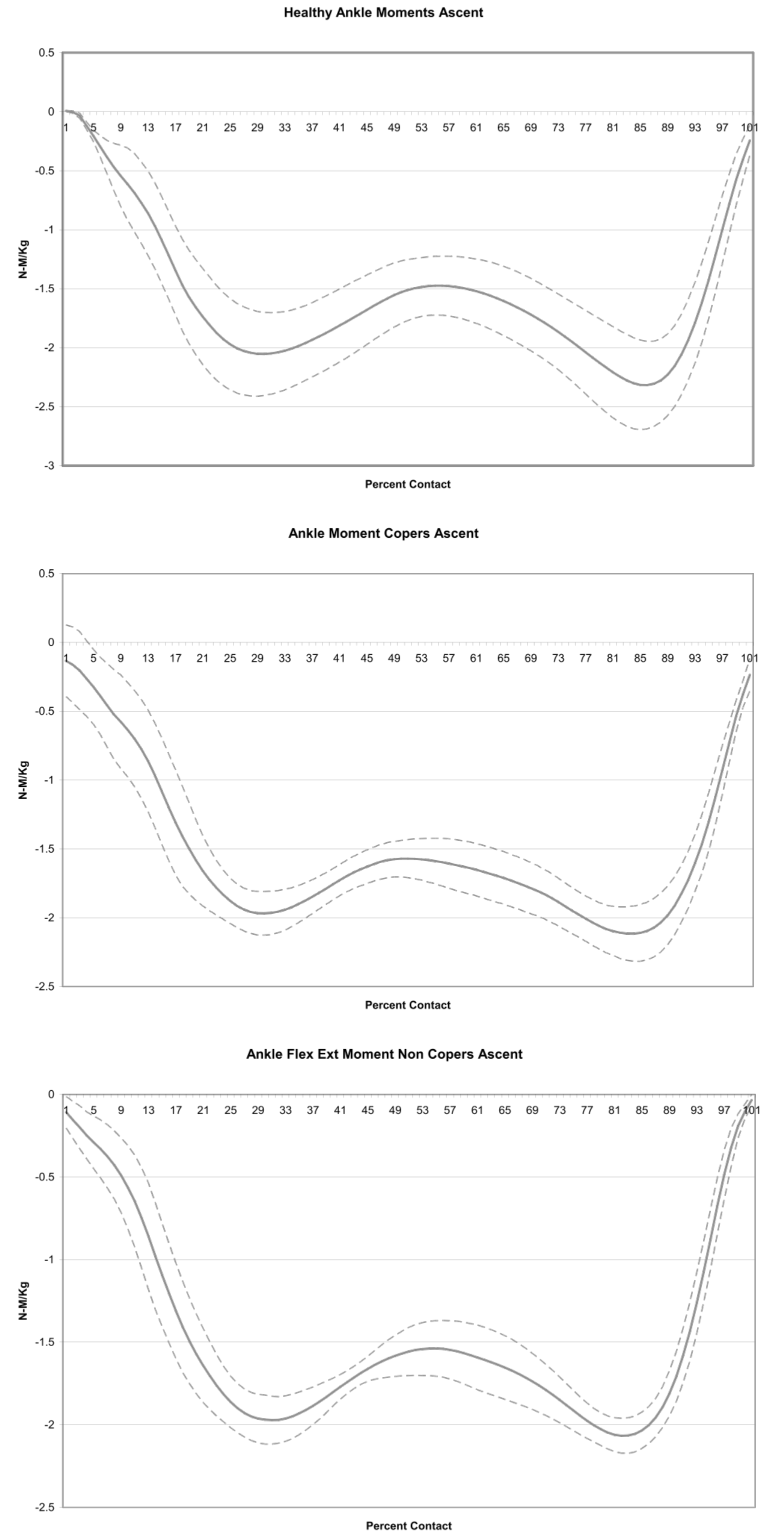


Figure 33: Ankle Flexion-Extension Moments of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

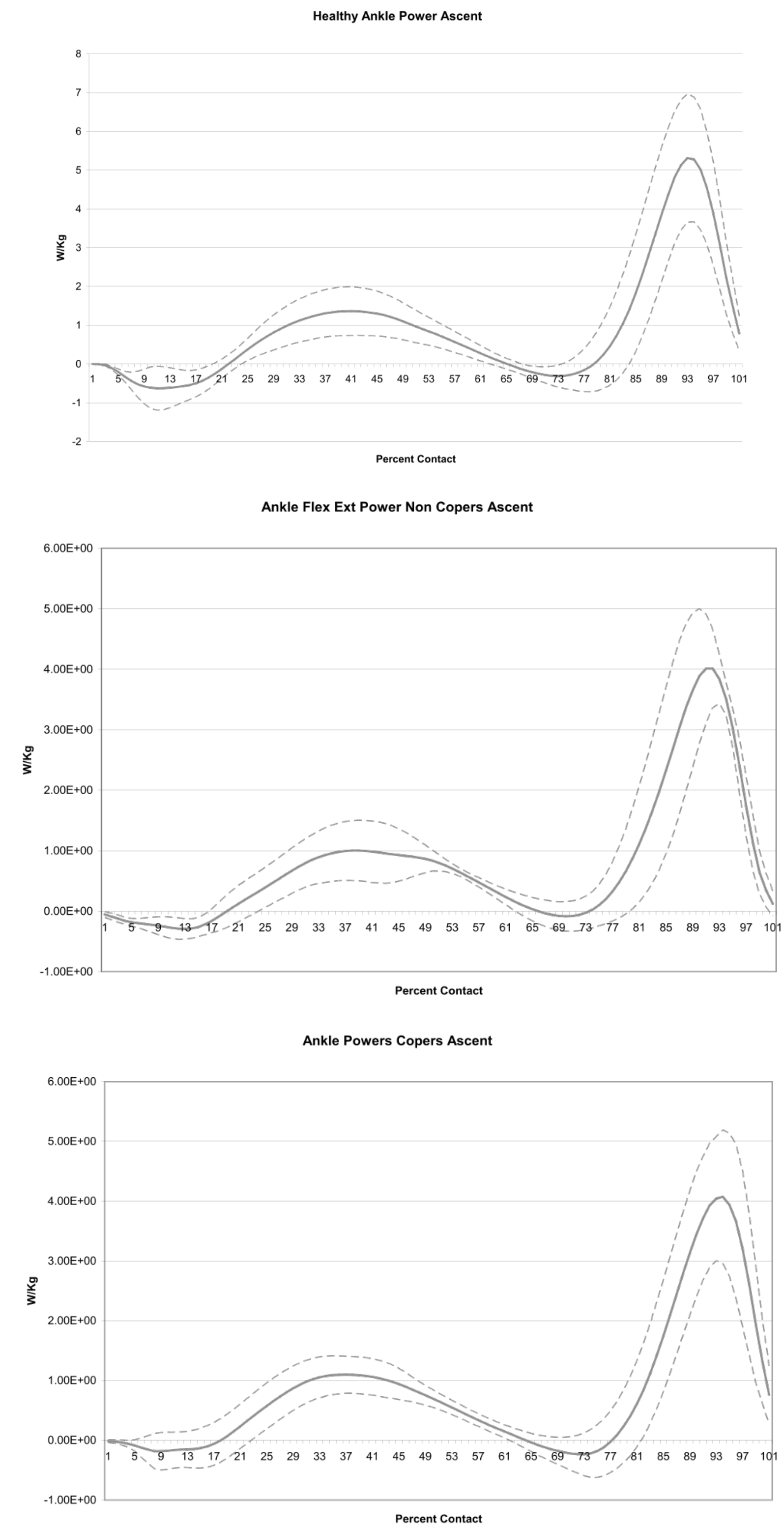


Figure 34: Ankle Power of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

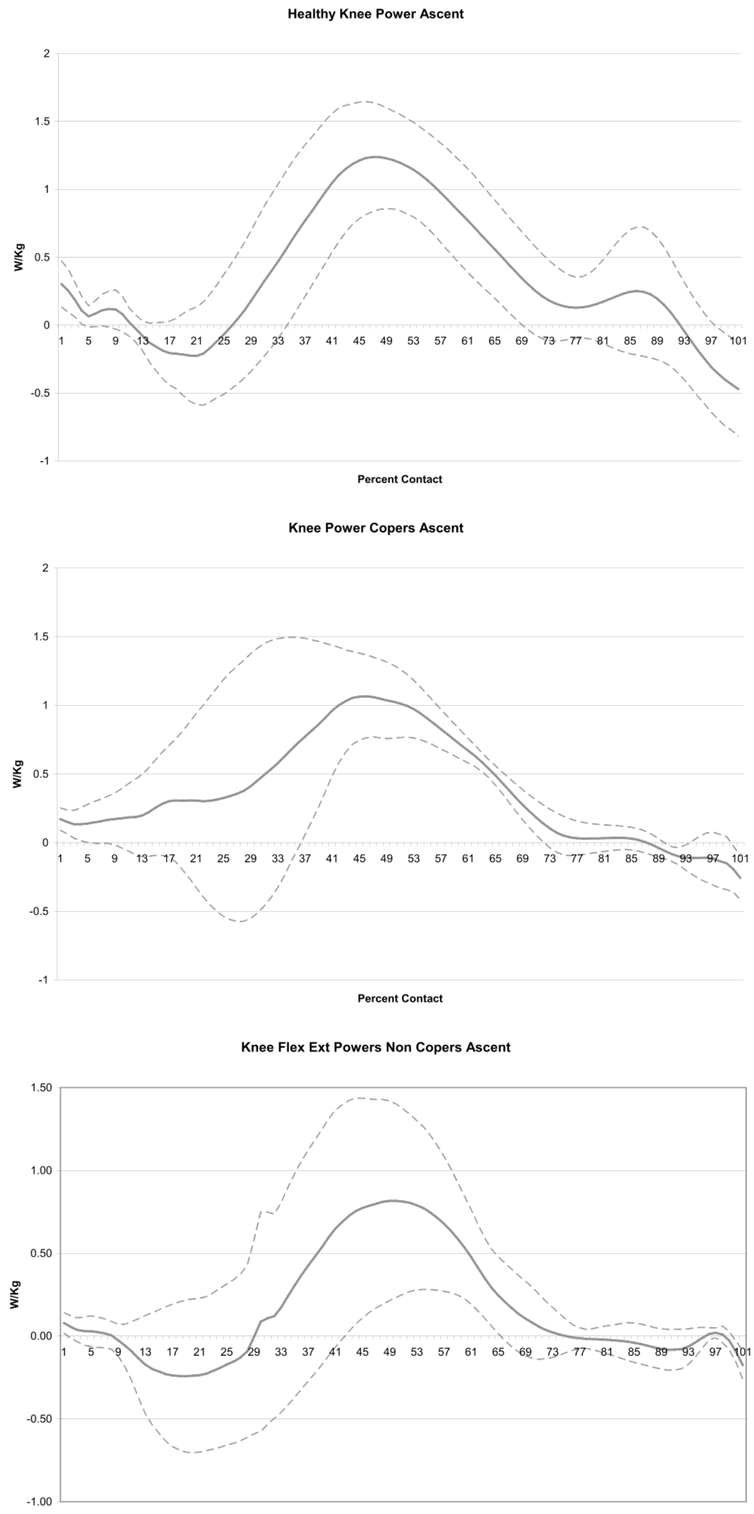


Figure 35: Knee Power of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

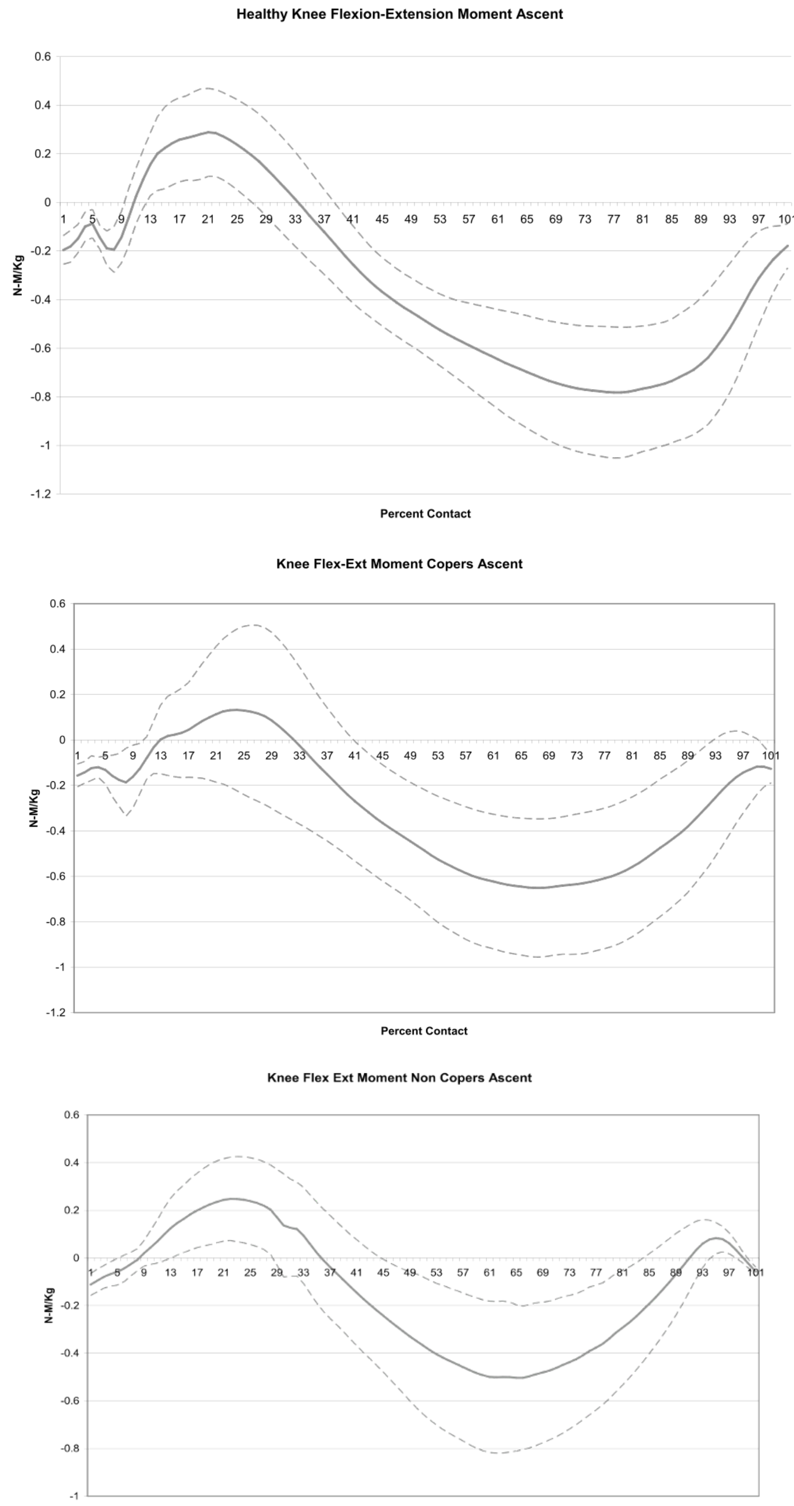


Figure 36: Knee Flexion-Extension Moment of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

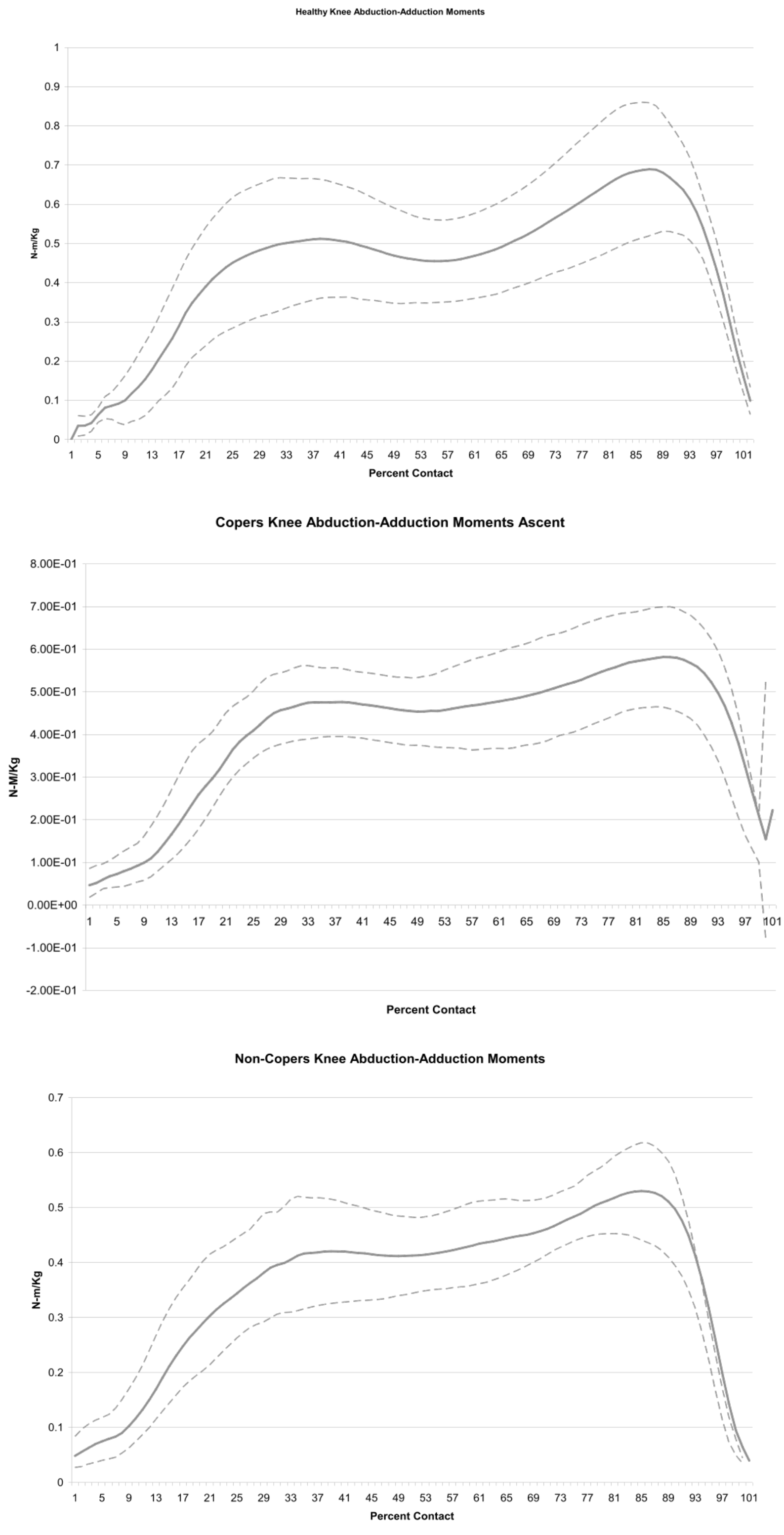
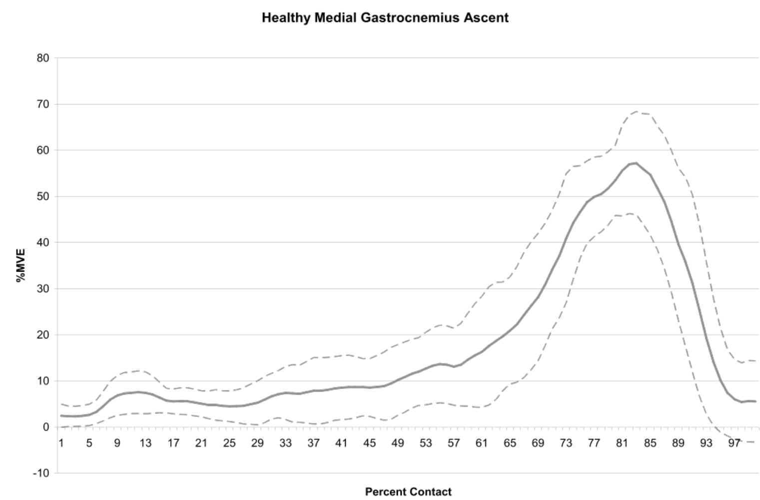
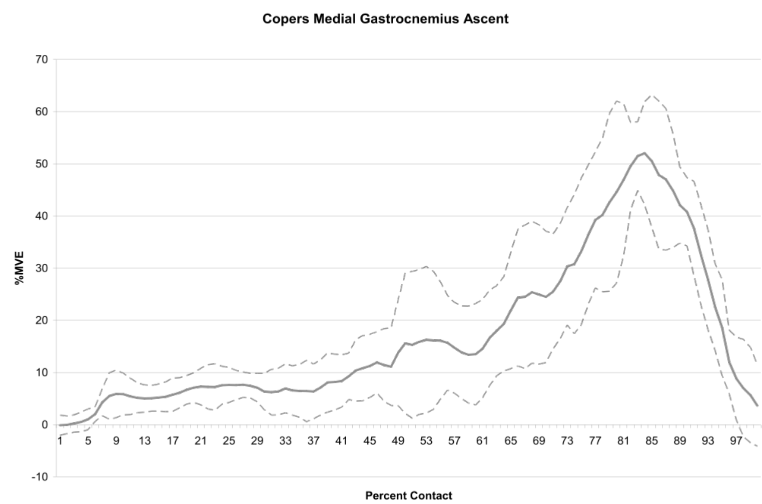


Figure 37: Knee Abduction-Adduction Moment of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)





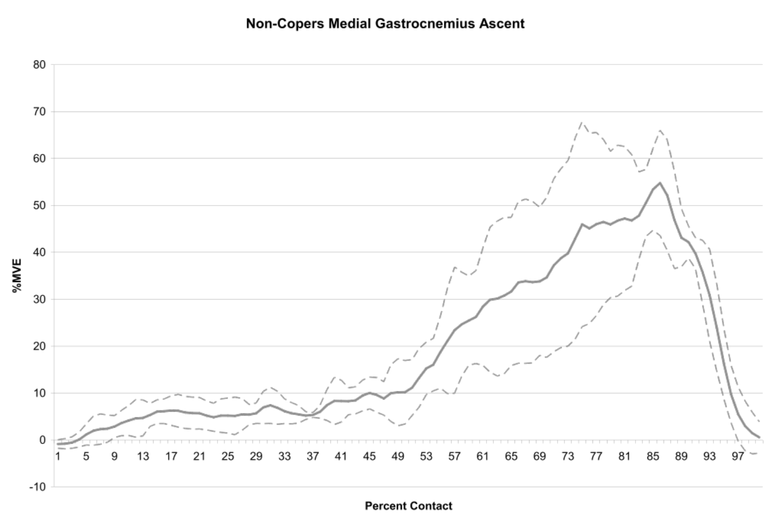


Figure 38: Medial Gastrocnemius Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

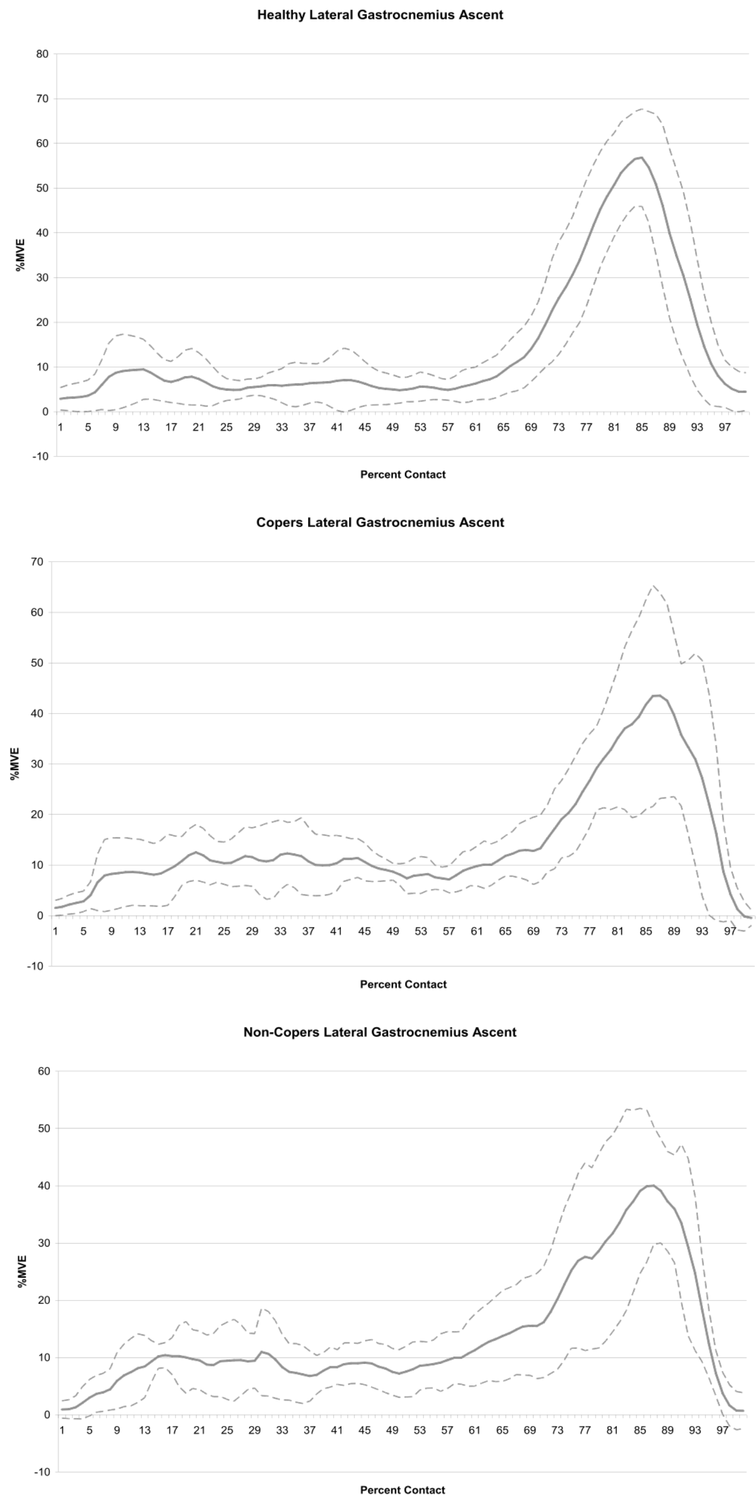


Figure 39: Lateral Gastrocnemius Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

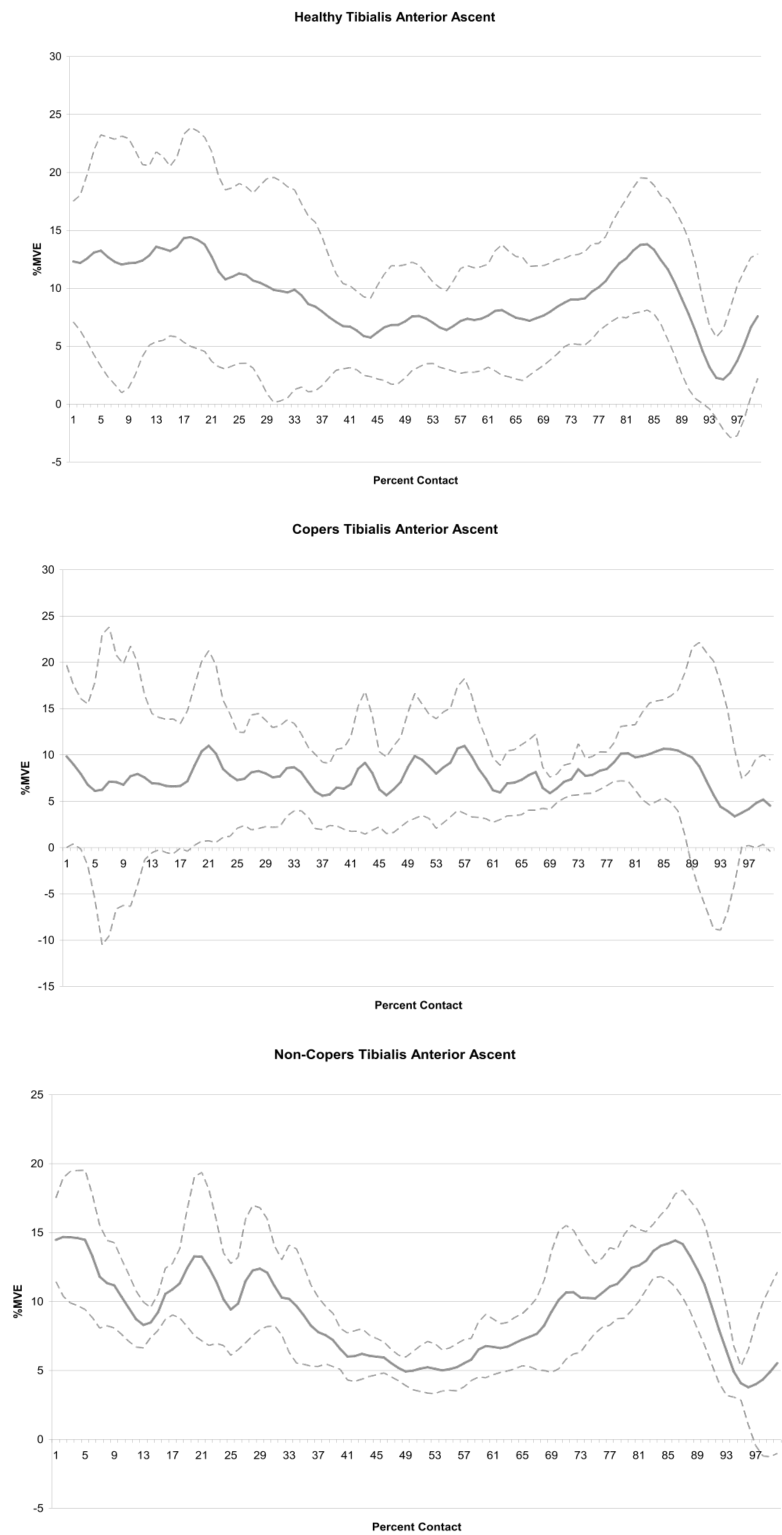


Figure 40: Tibialis Anterior Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

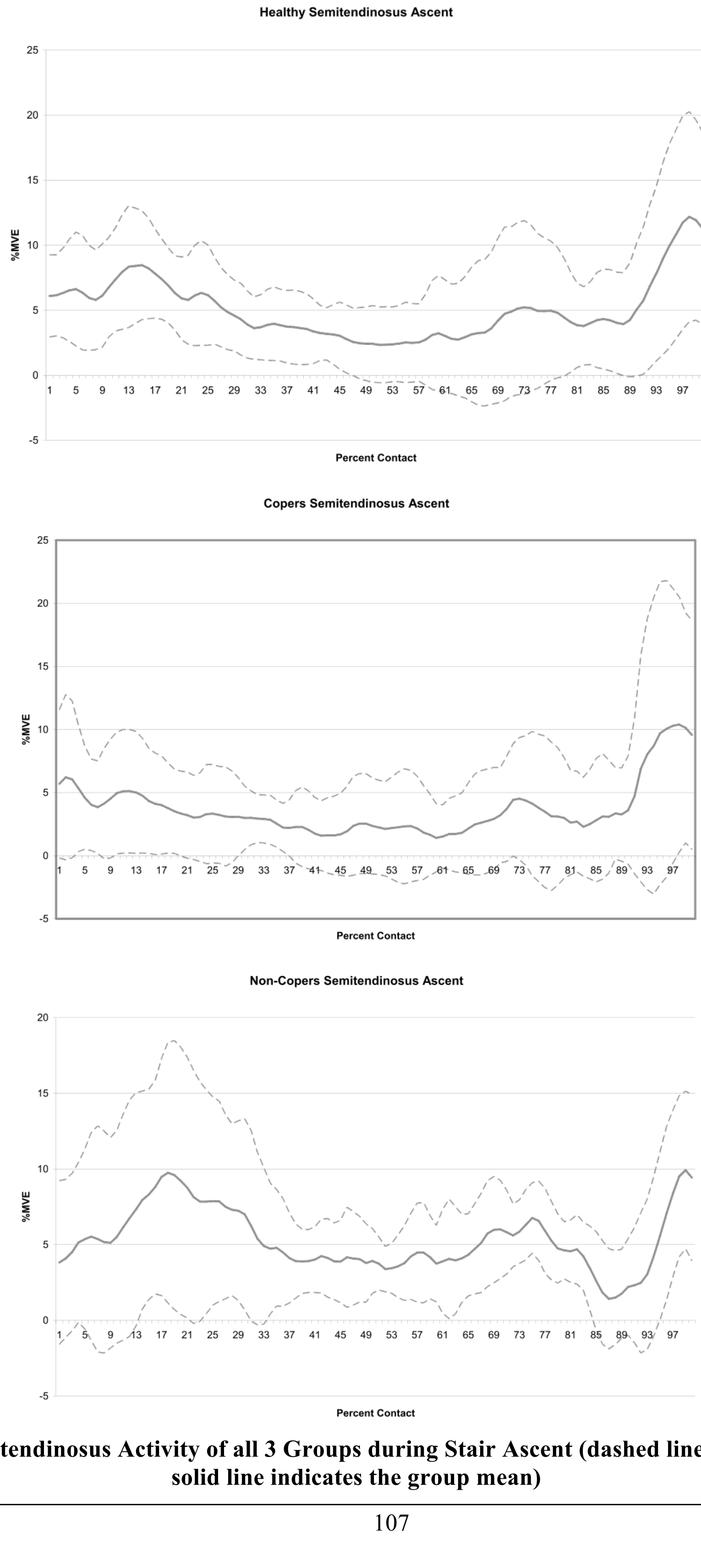


Figure 41: Semitendinosus Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

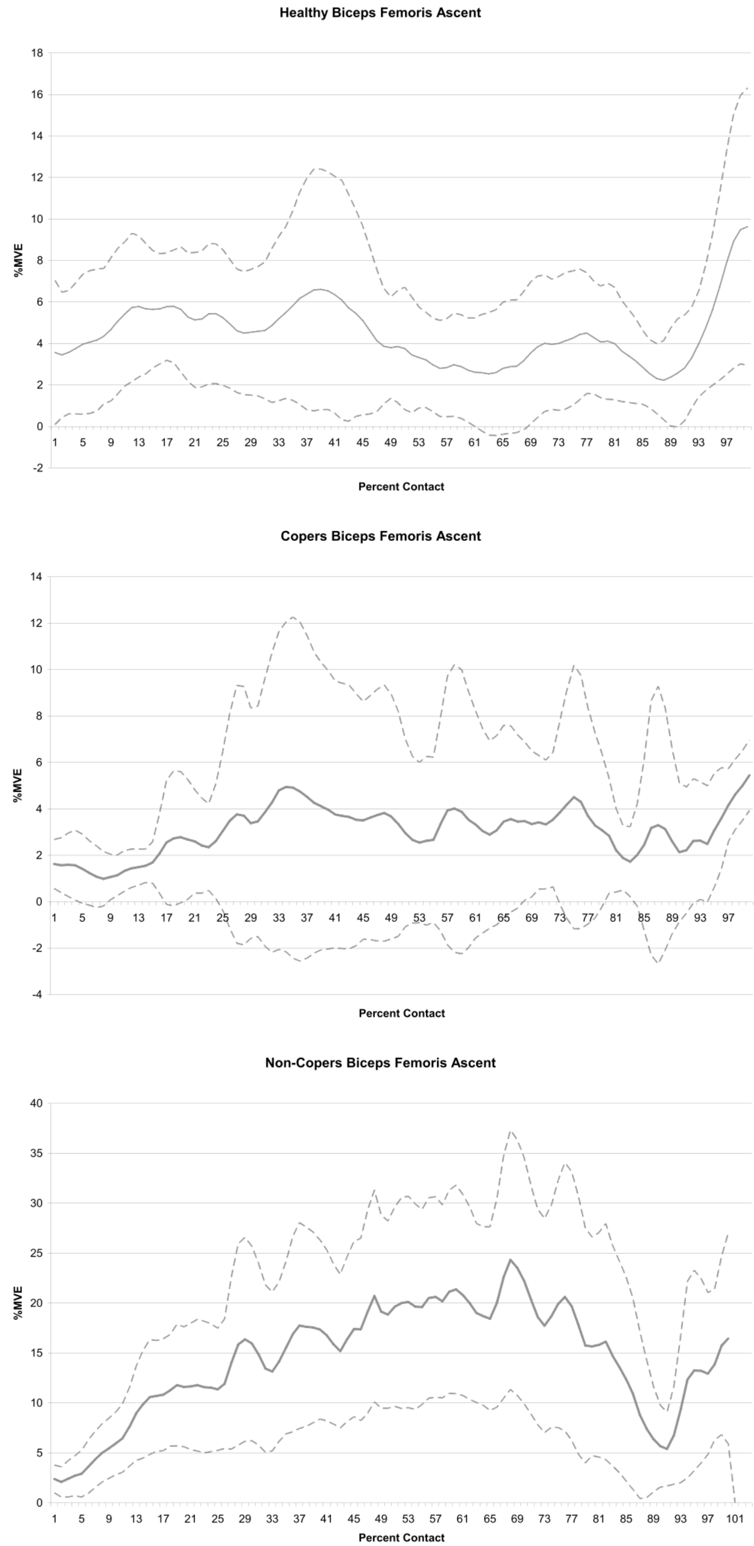


Figure 42: Biceps Femoris Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

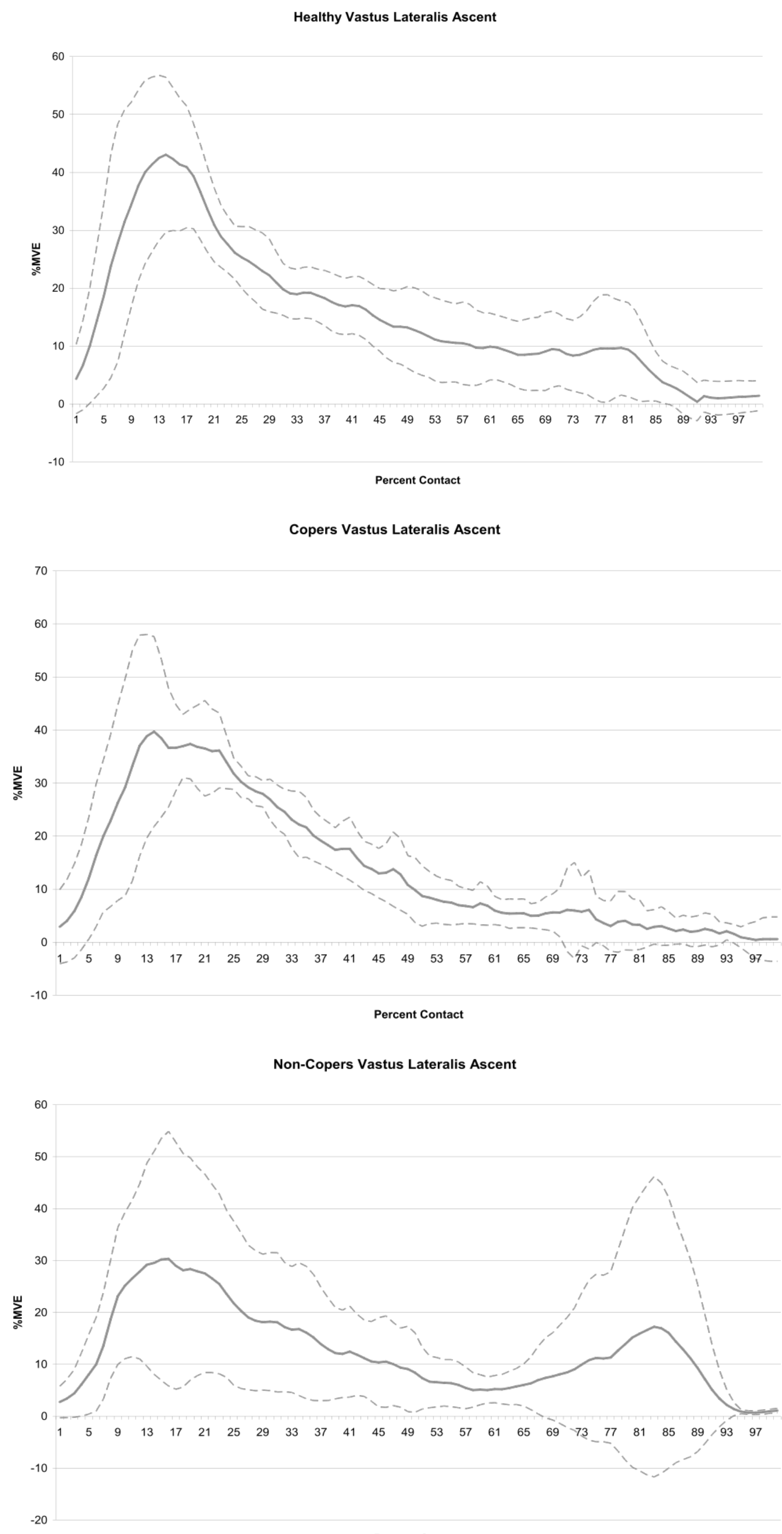


Figure 43: Vastus Lateralis Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

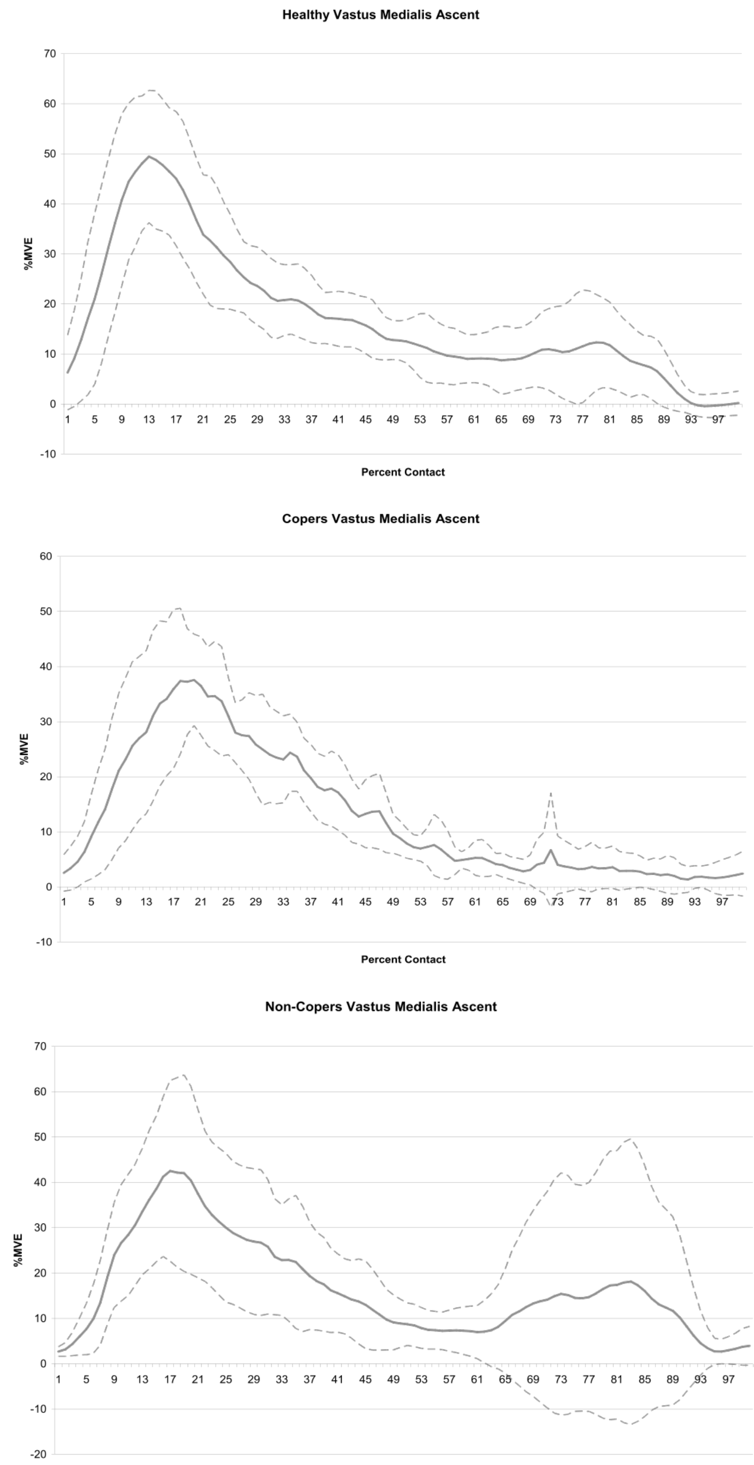


Figure 44: Vastus Medialis Activity of all 3 Groups during Stair Ascent (dashed lines indicate ±1 SD, solid line indicates the group mean)

Appendix 3: Descent Graphs

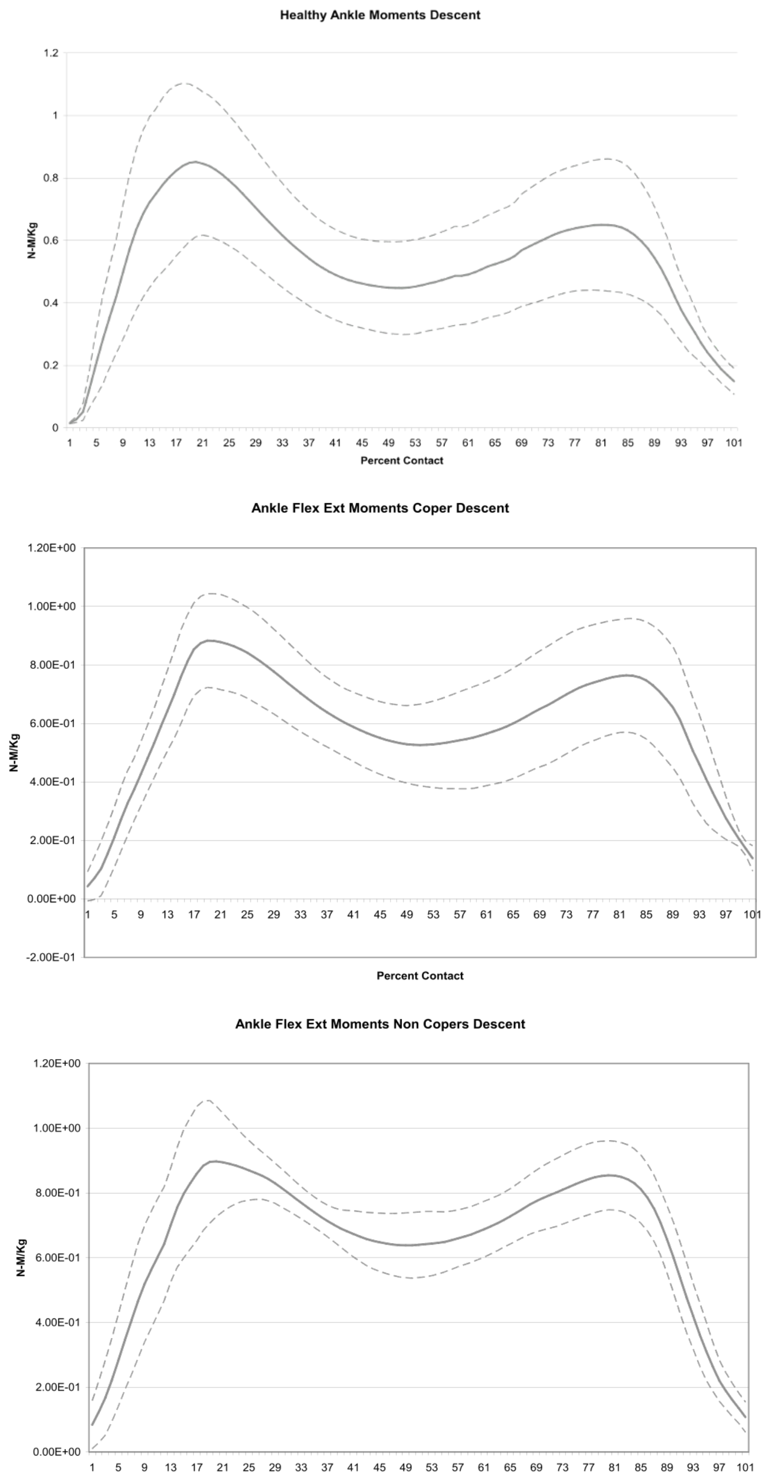


Figure 45: Ankle Flexion-Extension Moment of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

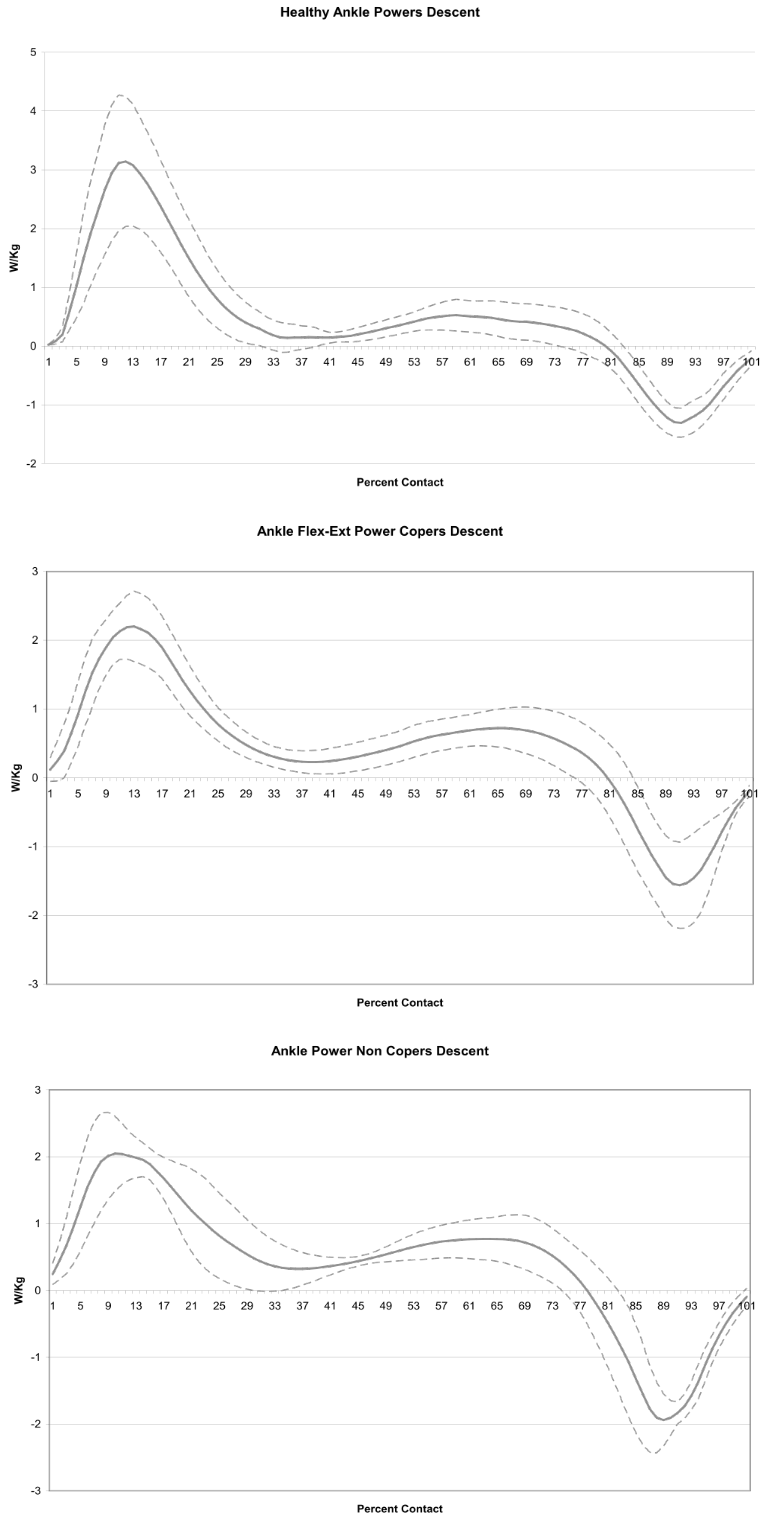


Figure 46: Ankle Power of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

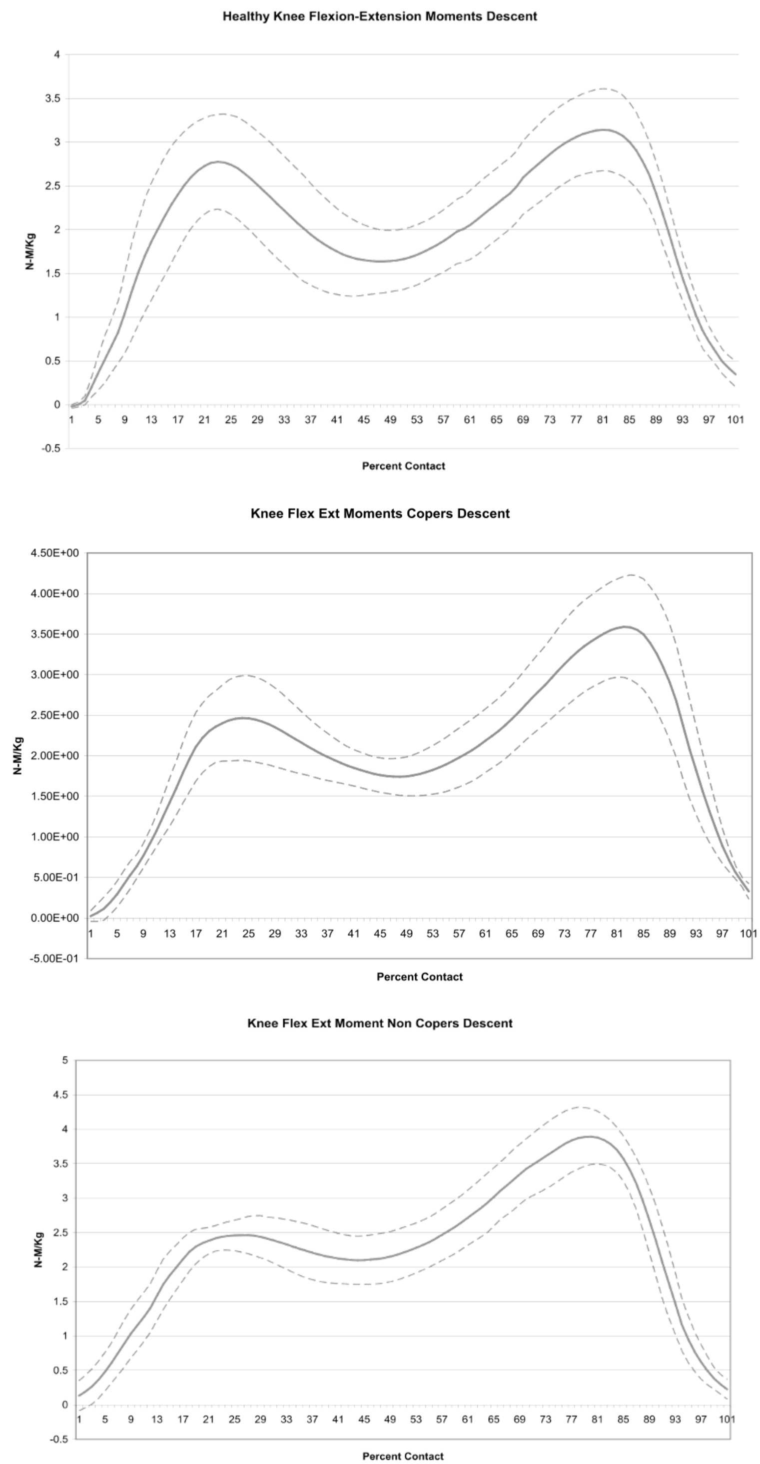


Figure 47: Knee Flexion-Extension Moment of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

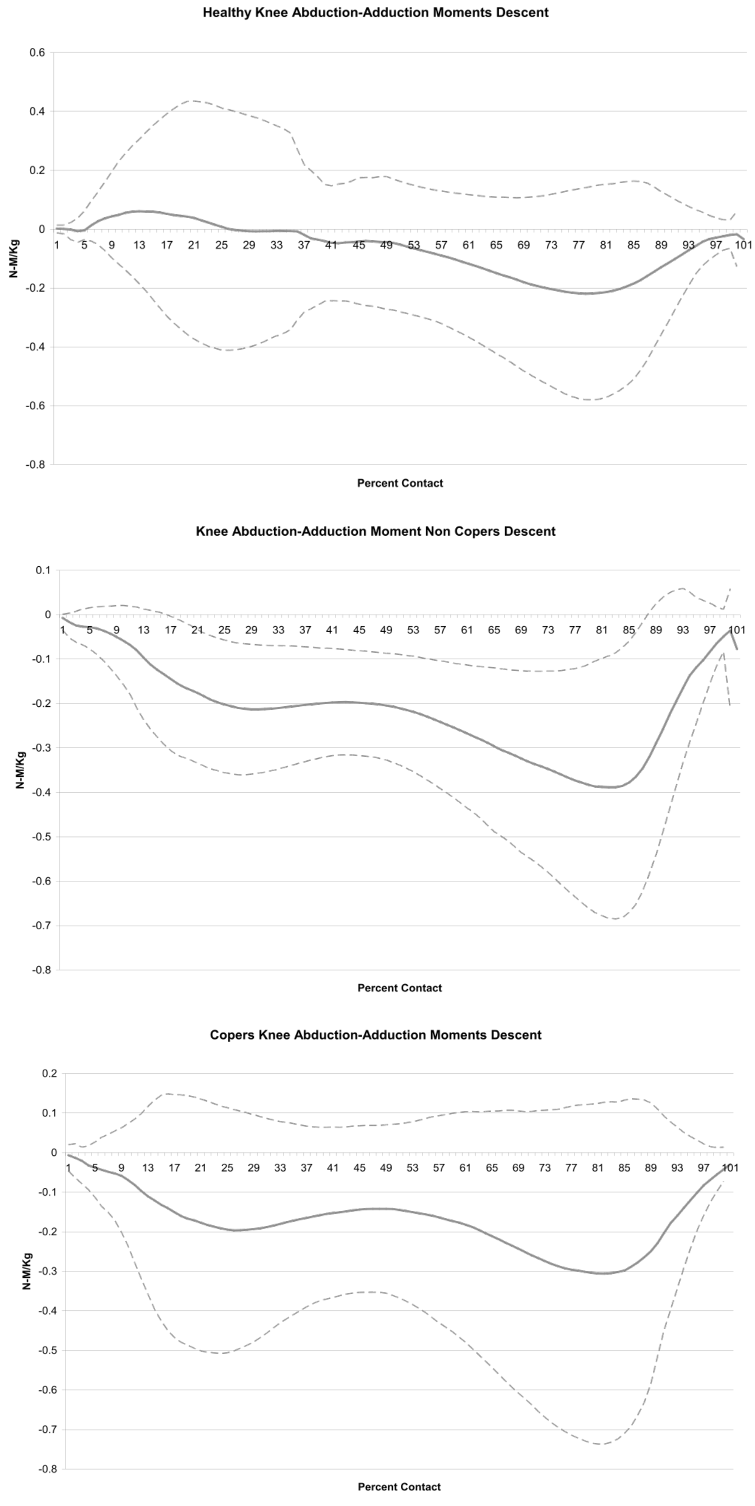


Figure 48: Knee Abduction-Adduction Moments of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

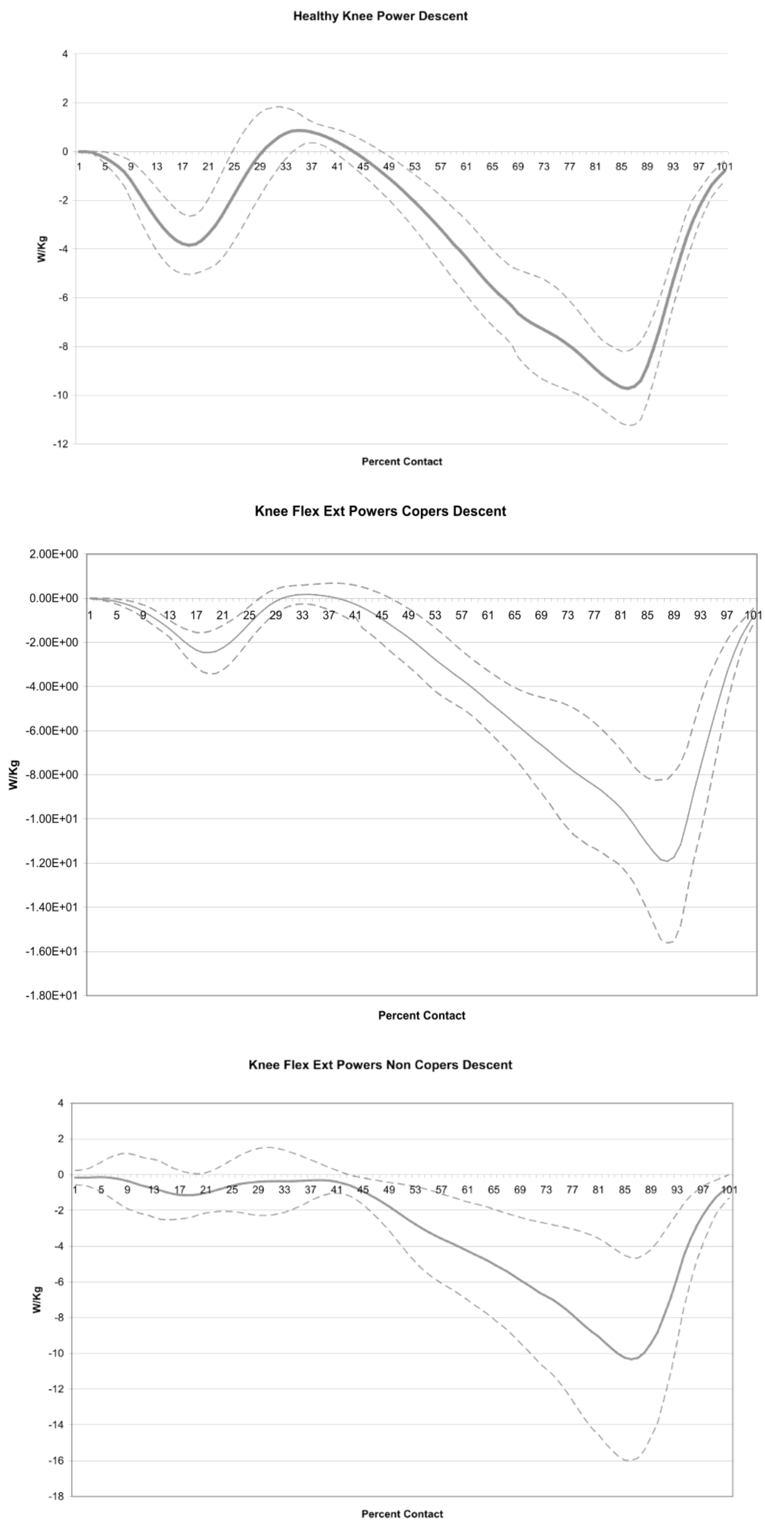


Figure 49: Knee Power of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

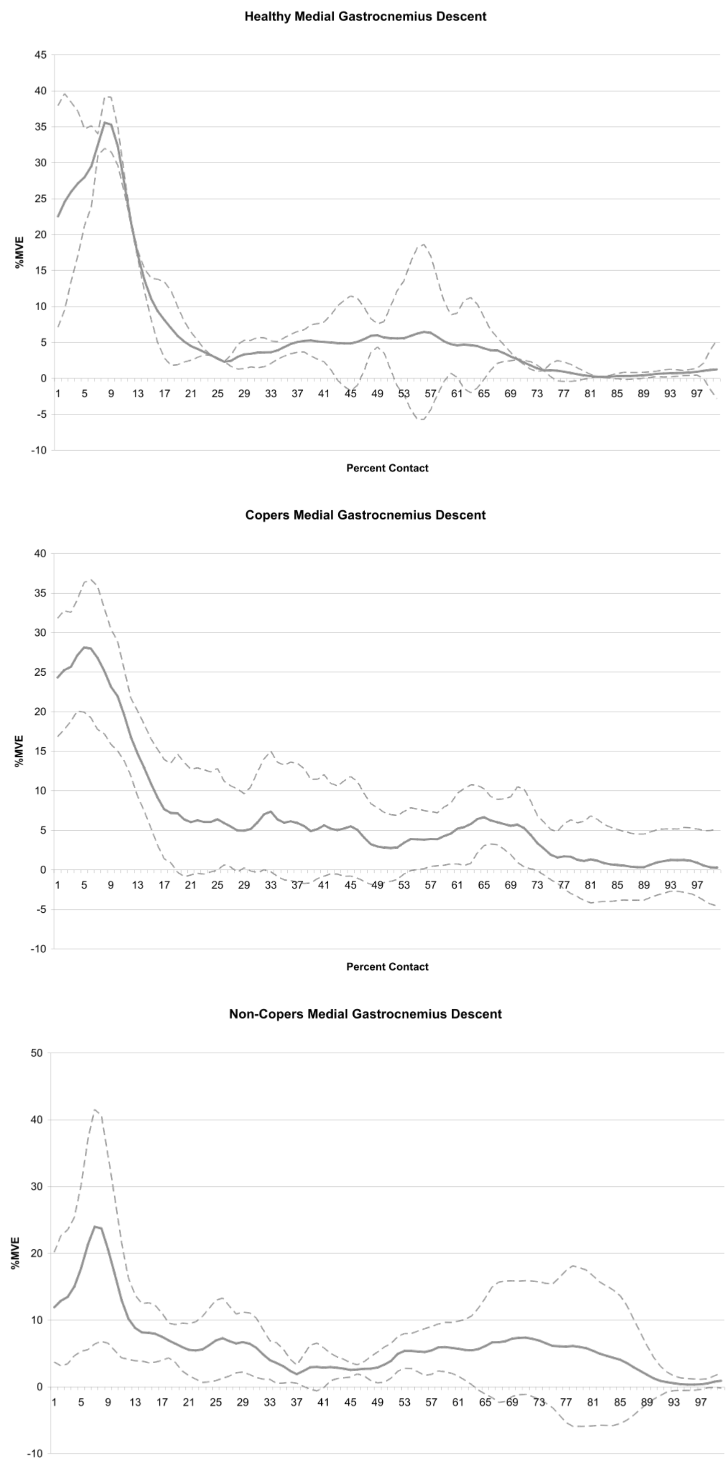


Figure 50: Medial Gastrocnemius Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

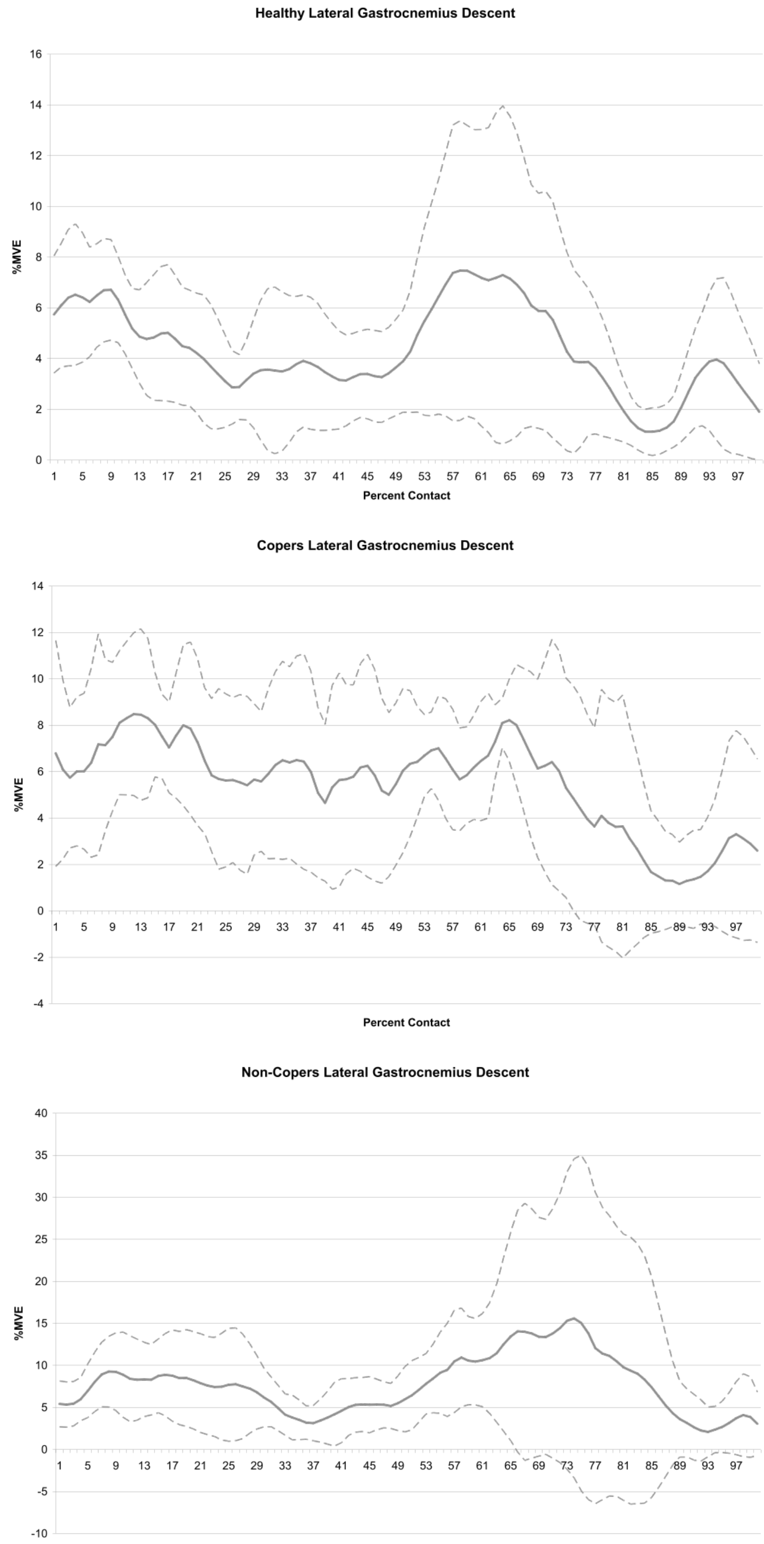


Figure 51: Lateral Gastrocnemius Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

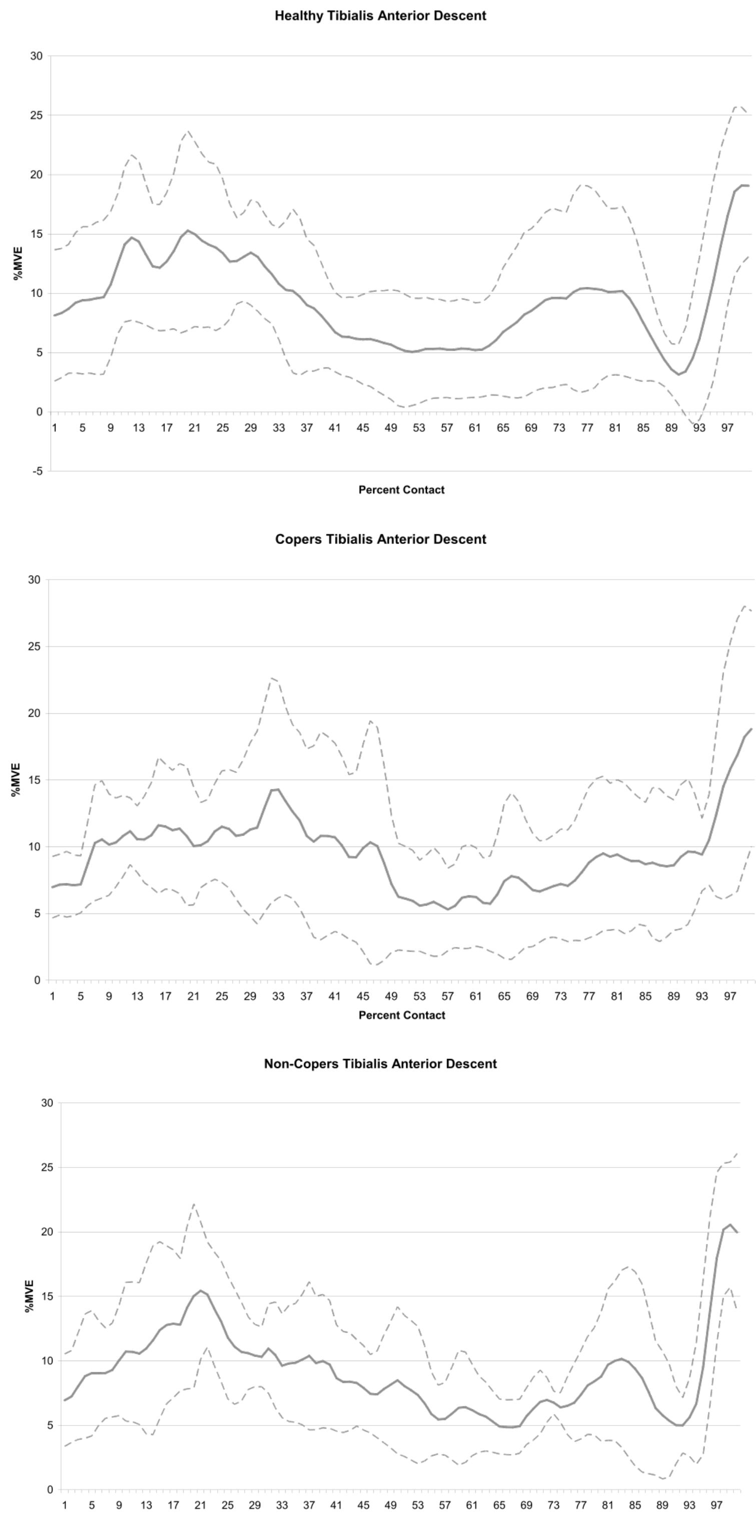


Figure 52: Tibialis Anterior Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

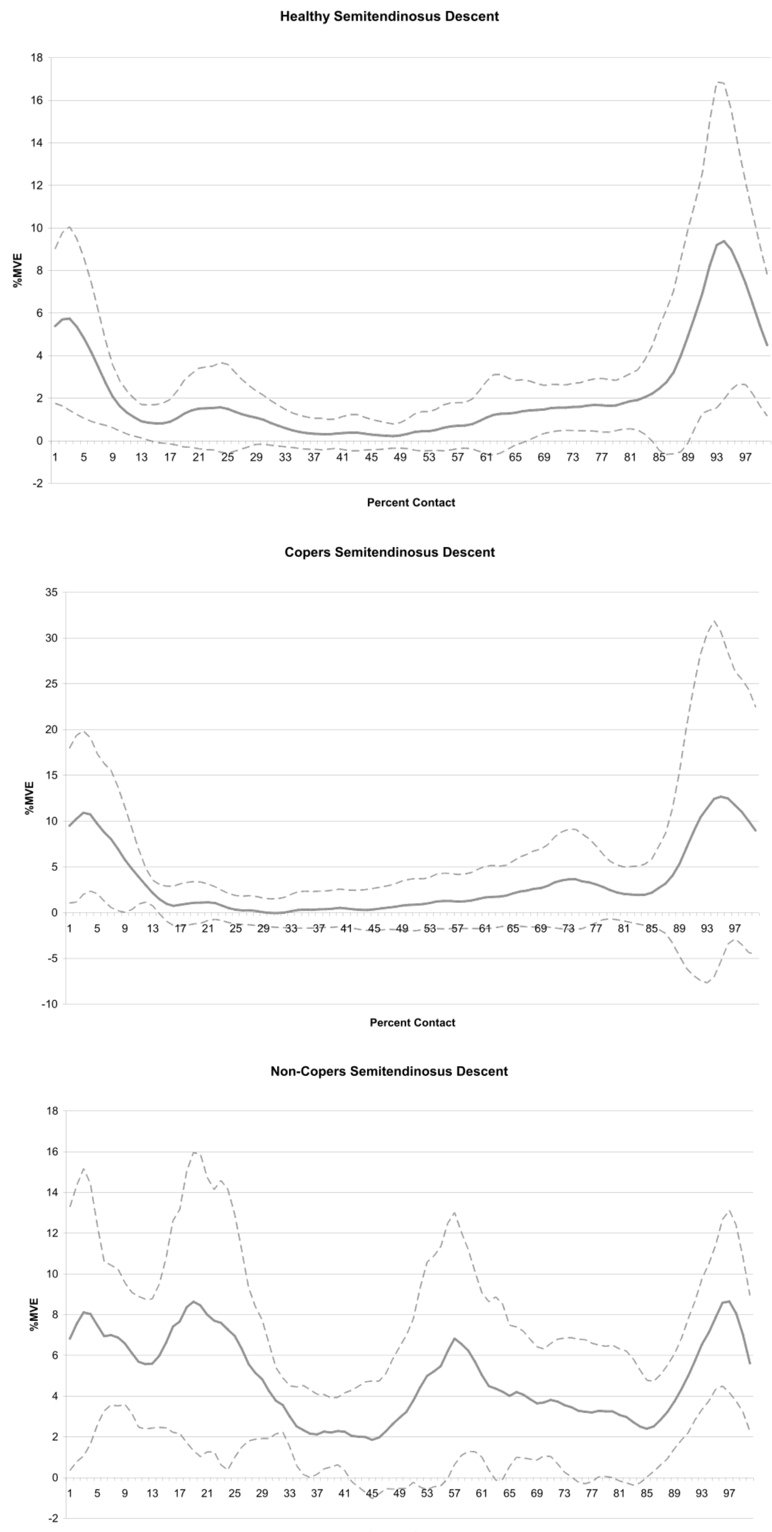


Figure 53: Semitendinosus Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

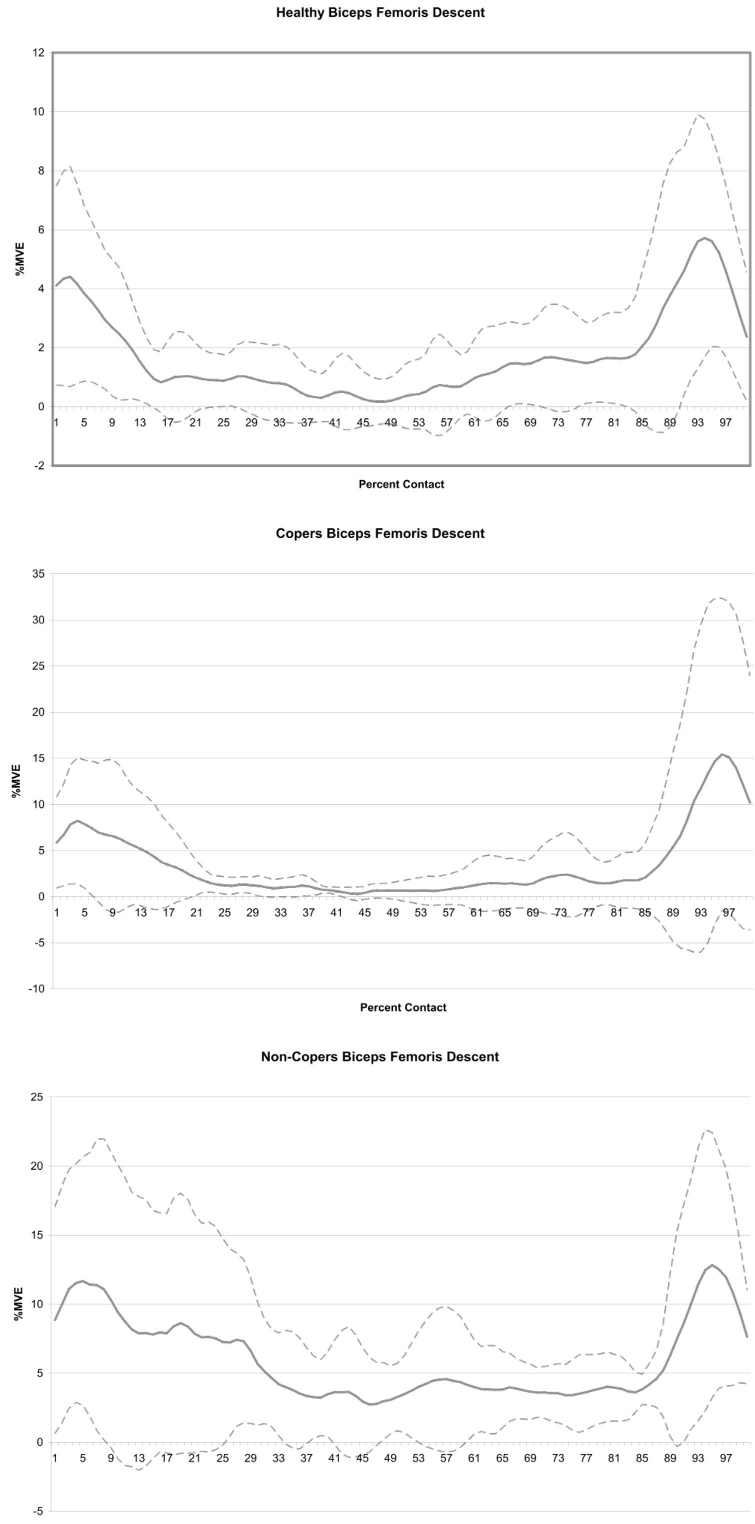


Figure 54: Biceps Femoris Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

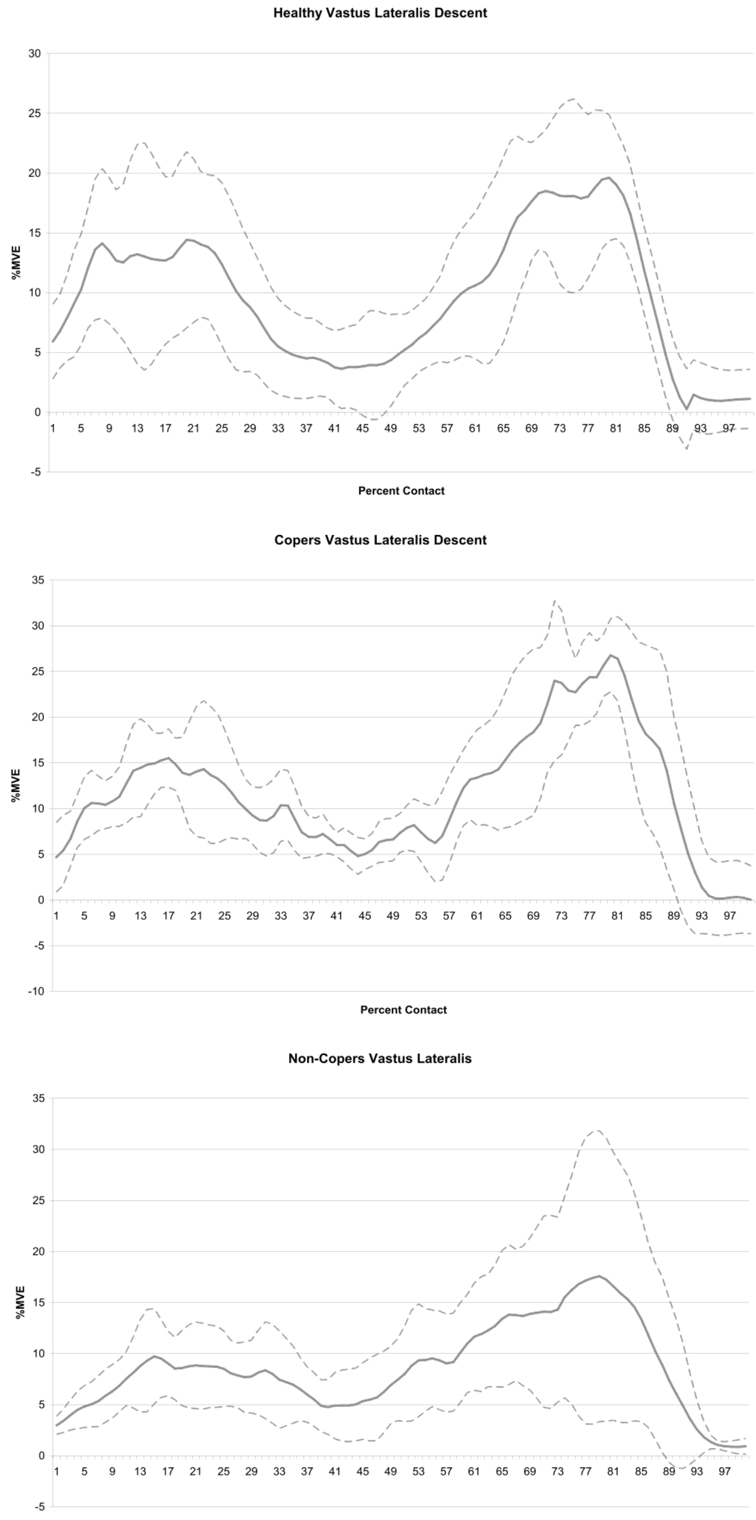


Figure 55: Vastus Lateralis Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

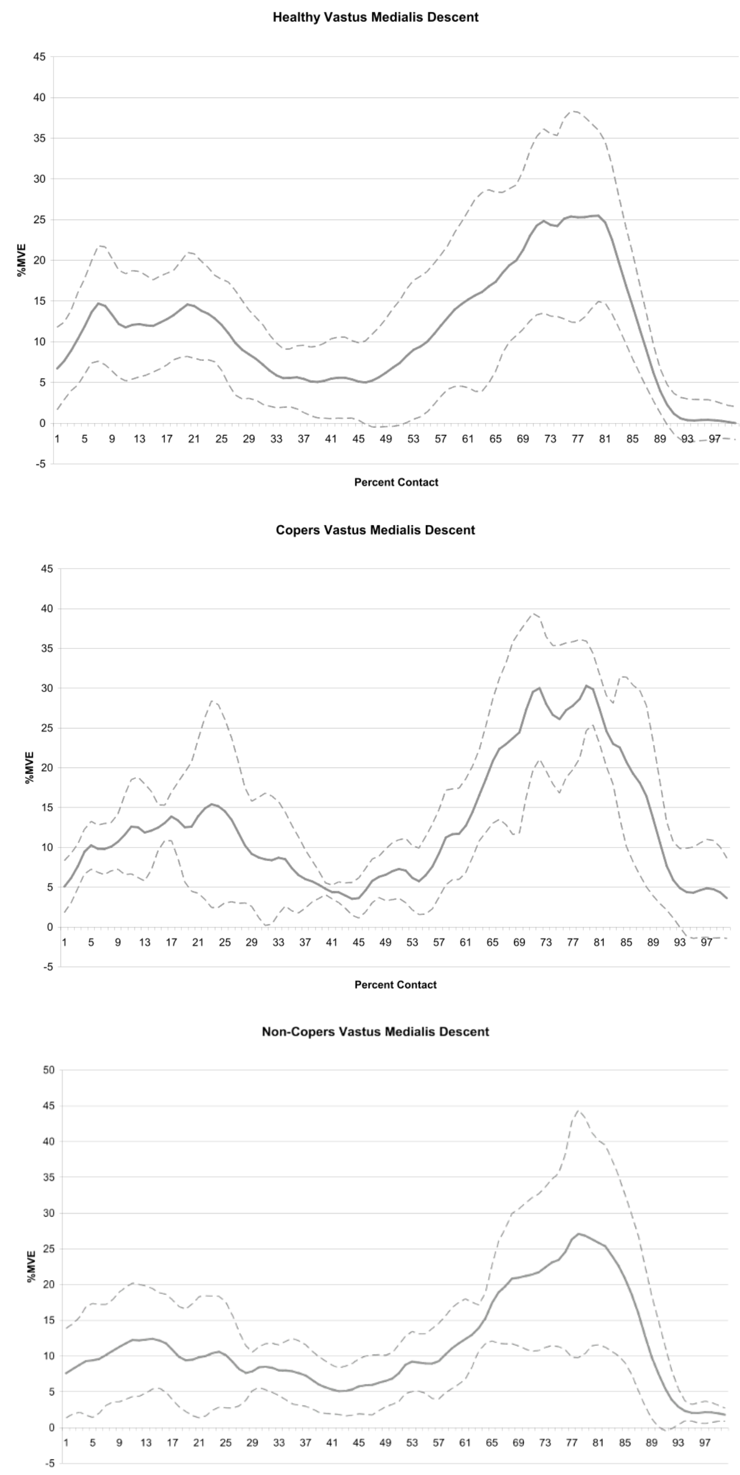


Figure 56: Vastus Medialis Activity of all 3 Groups during Stair Descent (dashed lines indicate ±1 SD, solid line indicates the group mean)

Appendix 4: F-values and P-values

Table 14: F-values and p-values for all EMG Variables

AEMG	F-value	p-value
Tibialis Anterior – Ascent	0.36	0.71
Medial Gastriocnemius- Ascent	0.85	0.45
Lateral Gastrocnemius – Ascent	0.89	0.43
Semitendinosus – Ascent	1.34	0.29
Biceps Femoris – Ascent	2.46	0.11
Vastus Lateralis – Ascent	1.52	0.25
Vastus Medialis – Ascent	0.27	0.71
Tibialis Anterior – Descent	0.09	0.91
Medial Gastriocnemius- Descent	0.26	0.78
Lateral Gastrocnemius – Descent	1.35	0.28
Semitendinosus – Descent	2.30	0.13
Biceps Femoris – Descent	2.99	0.08
Vastus Lateralis – Descent	1.08	0.36
Vastus Medialis – Descent	1.84	0.19

Table 15: F-values and p-values for the Kinetic Variables during Stair Ascent

	F-value	p-value
Knee Flex-Ext Moment – Early	0.75	0.49
Knee Flex-Ext Moment – Mid	2.34	0.13
Knee Flex-Ext Moment – Late	12.94	0.00
Knee Add-Abd Moment – Early	0.97	0.40
Knee Add-Abd Moment – Late	2.14	0.15
Ankle Moment – Early	0.30	0.74
Ankle Moment – Mid	0.40	0.68
Ankle Moment – Late	1.61	0.23
Knee Power – k1	0.23	0.79
Knee Power – k2	2.14	0.15
Knee Power – k3	0.30	0.75
Ankle Power – a1	3.18	0.07
Ankle Power – a2	0.87	0.44
Ankle Power – a3	0.45	0.65
Ankle Power – a4	0.75	0.49

Table 16: F-values and p-values for the Kinetic Variables during Stair Descent

	F-value	p-value
Knee Flex-Ext Moment – Early	1.36	0.28
Knee Flex-Ext Moment – Mid	4.89	0.02
Knee Flex-Ext Moment – Late	4.11	0.03
Knee Add-Abd Moment – Early	0.65	0.53
Knee Add-Abd Moment – Late	0.40	0.68
Ankle Moment – Early	0.07	0.93
Ankle Moment – Mid	4.19	0.03
Ankle Moment – Late	2.01	0.16
Knee Power – k1	4.70	0.02
Knee Power – k2	4.52	0.03
Knee Power – k3	0.99	0.39
Ankle Power – a1	3.10	0.08
Ankle Power – a2	4.14	0.03
Ankle Power – a3	1.48	0.26
Ankle Power – a4	1.36	0.28

Table 17: F-values and p-values for the EMG Variables during Stair Ascent

	F-value	p-value
Vastus Medialis – Peak Value	6.81	0.01
Vastus Medialis- % to peak	1.02	0.46
Medial Gastrocnemius – Peak Value	17.56	<0.001
Medial Gastrocnemius - % to peak	7.83	0.00
Biceps Femoris – Peak Value	0.14	1.00
Biceps Femoris - % to peak	1.18	0.37
Semitendinosus – Peak Value	0.49	0.84
Semitendinosus - % to peak	0.19	0.99
Lateral Gastrocnemius – Peak Value	4.08	0.01
Lateral Gastrocnemius - % to peak	5.96	0.06
Vastus Lateralis – Peak Value	1.59	0.20
Vastus Lateralis - % to peak	0.79	0.62
Tibialis Anterior – Peak Value	1.89	0.13
Tibialis Anterior - % to peak	0.31	0.95

Table 18: F-values and p-values for the EMG Variables during Stair Descent

	F-value	p-value
Vastus Medialis – Peak Value	0.12	0.89
Vastus Medialis- % to peak	0.38	0.69
Medial Gastrocnemius – Peak Value	2.39	0.12
Medial Gastrocnemius - % to peak	1.67	0.22
Biceps Femoris – Peak Value	2.54	0.11
Biceps Femoris - % to peak	0.97	0.40
Semitendinosus – Peak Value	1.20	0.34
Semitendinosus - % to peak	0.06	0.94
Lateral Gastrocnemius – Peak Value	1.39	0.28
Lateral Gastrocnemius - % to peak	0.36	0.70
Vastus Lateralis – Peak Value	3.97	0.04
Vastus Lateralis - % to peak	1.16	0.34
Tibialis Anterior – Peak Value	4.45	0.03
Tibialis Anterior - % to peak	0.18	0.84

Table 19: F-values and p-values for the Kinematic Variables during Stair Ascent

	F-value	p-value
Onset	1.18	0.33
25% Stance	1.17	0.34
50% Stance	0.96	0.40
75% Stance	1.21	0.32
Offset	1.11	0.38

Table 20: F-values and p-values for the Kinematic Variables during Stair Descent

	F-value	p-value
Onset	0.77	0.48
25% Stance	0.34	0.71
50% Stance	1.73	0.21
75% Stance	1.06	0.37
Offset	3.03	0.07