The Influence of Q-Angle and Gender on the Stair-Climbing Kinetics and Kinematics of the Knee

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 revisions, as accepted by my examiners.

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Abstract

Background: Knee joint motion and quadriceps activity play a crucial role in all lower limb tasks, especially those which are highly dynamic and weight-bearing. Due to anatomical differences between men and women such as height, leg length, and hip width, alignment and mechanics of the lower limb are different between males and females. An anatomical variable which is associated with alignment in the lower limb is the quadriceps muscle angle (q-angle). The purpose of this study is to determine if there is a relationship between q-angle, activity of the quadriceps and hamstring muscles and the kinetics and kinematics of the knee during stair-climbing. An investigation on the reliability of q-angle measurements was also made prior to the primary study.

Methods: To test the interclass reliability of q-angle measurements, three individuals measured the q-angle on 20 subjects. The primary researcher measured the same twenty individuals on three separate days to determine intra-rater reliability.
The primary study involved 10 male and 10 female subjects completing 20 stair-climbing trials (10 ascent, 10 descent). Kinematic and kinetic data were collected on the lower limbs as well as electromyography (EMG) on two quadriceps muscles and one hamstring muscle. Knee joint peak and occurrence of peak moments, average EMG amplitude and peak and occurrence of peak EMG were analyzed by gender and high and low q-angle. A two way analysis of variance (ANOVA) was used to test the statistical significance of each measured variable ($\alpha = 0.05$).

Results & Discussion: The inter-rater reliability for q-angle was low (0.27-0.78) but the intra-rater reliability showed q-angle measurements to be very reliable (0.80-0.95). For study 2, it was found that females had increased vastus lateralis and vastus medialis peak EMG and average EMG amplitudes for stair ascent and descent compared to males. Furthermore, for descent only, females demonstrated having delayed occurrence of peak EMG for vastus lateralis and biceps femoris, and exhibited an increased peak knee extension moment and a decreased peak knee adduction moment compared to males. For q-angle, there was a significant difference found for biceps femoris occurrence of peak EMG during descent, with the high q-angle group having delayed occurrence of peak. For ascent, the high q-angle group had significantly increased average vastus lateralis EMG and an earlier occurrence of knee abduction moment. Q-angles were found to be higher for women compared to men.

Conclusion: This study confirms that gender differences do exist in knee moment and thigh EMG parameters with stair ascent and stair descent. With the high incidence of significant findings for the quadriceps muscle, further investigation is warranted to determine if a relationship does exist between q-angle and knee joint function. It would also be recommended that hip mechanics be included in future studies due to the difference seen in adduction moments at the knee.

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Table of Contents

List of Tables

List of Figures

List of Appendices

1. Introduction

Women have an increased prevalence of tibio-femoral osteoarthritis (OA) of the knee as well as an increased risk of developing the disease (Kaufman et al., 2001; Messier et al., 1992; Hunter et al., 2005; Zhang et al., 2004; Hughes et al., 2000). Research to date has not been able to isolate specific causes as to why this gender difference occurs. A factor found to play a significant role in the advancement of OA in general is knee alignment (Cerejo et al., 2002). An anatomical variable which is associated with alignment in the lower limb is the quadriceps muscle angle (q-angle). The q-angle is the angle between a vector connecting the anterior superior iliac spine (ASIS) to the patella (knee cap) and a vector connecting the patella to the tibial tuberosity (Livingston and Spaulding, 2002). The first vector represents the quadriceps muscles and the second vector represents the patellar tendon. It has statistically been shown that the q-angle of females is larger than that of males (Woodland and Francis, 1992). In numerous studies, women have exhibited excessive q-angle measures. Recently, values between 8° - 10° for men and up to 15° for women are considered normal and that values above this can be problematic (Livingston and Spaulding, 2002; Greene et al., 2001; Livingston, 1998; Boden et al., 1997). The majority of what is reported in the literature is an angle in excess than $\geq$15-20 is considered to be problematic (Horton and Hall, 1989; Livingston, 1998; Greene et al., 2001; Byl et al., 2000). Lathinghouse, *et al.*, (2000) reported that an excessive q-angle may predispose women to greater lateral displacement of the patella during activity requiring high levels of quadriceps activation. But the connection of these variables to the mechanics of the knee in women has not yet been made.

In addition to structural factors, functional differences between genders have also been identified. Women have exhibited increased knee flexion angles during stair ascent (Hughes et al., 2000). Women also have an increased extensor moment during stair-climbing and walking (Kaufman et al., 2001), which would increase loading at the knee. Abnormal loading on a continuous basis or in frequent intervals cause injury and additional wear and tear on a joint (Luepongsak et al., 2002). Anatomical differences are one possible explanation for this gender disparity which will be investigated here using the quadriceps muscle angle. There is evidence that the q-angle increases with increased knee flexion angles (Livingston and Spaulding, 1999). The increased angle would pull on the tibia which could affect the tibio-femoral contact during movement. It has also been reported that with decreased q-angles, the valgus orientation of the knee decreases (Mizuno et al., 2001). Decreased q-angles also exhibited increased peak torques at the knee (Byl et al., 2000).

Stair-climbing has proven to be an effective method of testing lower limb joint function as it stresses the mechanics of the knee and also represents a common activity of daily living. External adduction moments, which are related to alignment, were found to be increased 8-11 fold in stair-climbing and walking compared to standing (Luepongsak et al., 2002). This will change the contact area within the knee joint, distributing the forces onto a different area within the knee joint compared to walking. With the increased loading, joint range and muscle activity that occur during stair-climbing compared to walking, this proves to be an excellent method for analyzing knee mechanics and q-angle. Connecting the aforementioned functional and structural variables to the existent gender difference of q-angle during stair-climbing may help to expose trends between genders.

1.1 Statement of Purpose

The purpose of this study is to determine if there is a relationship between the following variables: (i) the magnitude of the quadriceps angle, (ii) the activity of the quadriceps and hamstring muscles and (iii) the kinetics and kinematics of the knee during stair ascent and descent. The secondary purpose is to: (i) test the intra and inter reliability of the q-angle measurement.

1.2 Problem Investigated

Is the quadriceps muscle angle (q-angle) a reliable measurement between and within testers? Are there differences in knee joint kinetics during stair-climbing between genders and q-angles?

1.3 Statement of Hypotheses

It is hypothesized that women and men with a larger quadriceps angle will have (i) increased quadriceps activity and (ii) increased peak flexion/extension moments at the knee. It is further hypothesized that, when comparing men and women alone: (iii) women will have increased muscle activity, and (iv) increased knee extension and abduction moments.

2. Review of Literature

The following review of literature will investigate research surrounding the kinematics and kinetics of the knee as well as muscle activation and strength patterns. These components will be looked at individually relating to knee osteoarthritis, stair-climbing and q-angle.

2.1 Knee Osteoarthritis Risk Factors

Osteoarthritis is the most common form of arthritis, which is caused by the breakdown of cartilage, causing pain and inflammation in the joint (Kaufman et al., 2001). Approximately 1 in 10 Canadians suffer from osteoarthritis, which incurs $4.4 billion in health care costs per year (Siler et al., 1997). Osteoarthritis can affect any joint but usually affects hips, knees, hands and the spine (Siler et al., 1997). The disease is particularly debilitating when it affects the joints of the lower limb. When gait is affected by disease or injury, it greatly affects a person's ability to function and their quality of life. In a study conducted by Guccione et al. (1994), it was determined that knee osteoarthritis is a leading cause of functional limitation in an elderly population next to stroke. However, an understanding of the mechanisms that cause osteoarthritis of the knee is still lacking.

Studies involving osteoarthritis patients have attempted to increase the applicability of the findings by also testing healthy subjects to have as a comparison. One study had three subject groups composed of elderly with OA, healthy elderly and healthy young respectively complete three activities of daily living (Hortobagyi et al., 2005). It was found that people with OA had higher hamstring and quadriceps co-activation than the healthy, but the healthy elderly also had higher coactivity than the young (Hortobagyi et al., 2005). Additional irregularities in knee function have been reported. It has been reported that arthritic patients had decreased knee range of motion (ROM), reduced isometric strength and lower ground reaction forces (Stauffer et al., 1977; Gyory et al., 1976). Messier *et al.,* (1992) found similar ROM and strength decreases with the diseased population but did not find a significant difference in the ground reaction forces. They did, however, find a decreased loading rate of the affected knee (Messier et al., 1992). The muscular weakness found in OA patients should cause decreased control of the knee which would affect how the leg contacts the ground during movement, causing an increased loading force at the knee. However, the increased coactivity which was stated above is compensating for the muscular weakness. With the decreased ground reaction force, the force sustained by the knee should be reduced which would minimize the pain experienced during activity, again acting as a compensatory mechanism. A decreased knee extensor moment has been seen in OA patients, which is thought to be a subconscious effort to reduce the forces at the knee and reduce the pain (Kaufman et al., 2001). But this again brings forth the question as to whether the moment is reduced due to the muscular weakness or is the weakness caused by a lack of use of the muscles to offset the pain.

The muscles are only one component of the knee as a whole. Injuries to the knee joint which damage the cartilage, bone or ligaments have also been implicated as a risk factors for knee osteoarthritis. One of the only prospective cohort studies completed to date in this area studied more than 1300 subjects over a thirty-five year period and determined that significant trauma to the knee occurring before the initial collection and during the follow-up period had substantially increased the risk of developing knee or hip OA (Gelber et al., 2000). A study correlating pain to knee function found that patients with pain from knee OA had reduced quadriceps function and activation patterns, but following the administration of an anesthetic and placebo into the knee joint, an increase in quadriceps MVC (maximum voluntary contraction) was seen (Hassan et al., 2002). With the placebo having a similar effect to the drug, this could be demonstrating a protective psychological tendency which could be causing the decreased quadriceps strength in the initial stages of the disease.

It is known that osteoarthritis can initially affect only one compartment of the knee in the beginning stages of the disease. Knee alignment may account for a major part of this. Varus alignment is associated with medial OA and valgus with lateral. There was found to be a significant risk increase for a medial OA related to a varus alignment with OA grade 2 and 3 knees (Cerejo et al., 2002). There was a relationship with valgus alignment and the lateral compartment, but was less significant (Cerejo et al., 2002). However, malalignment does not necessarily indicate a cause. The varus/valgus deformity could be caused by disease progression. This can be explained by the fact that in normally aligned knees, the medial side is loaded disproportionately compared to the lateral (Morrison, 1970). A radiographic study, taking into account lifetime activity, also looked at the prevalence of medial and lateral compartment OA (Zhang et al., 2004). The effect of occupational squatting was found to increase development of lateral OA in men and both medial and lateral OA in women (Zhang et al., 2004). Another study involving gender differences found that women had significantly greater knee extension moments which increased knee loading (Kaufman et al., 2001). The increase in knee loading helps to justify the increased prevalence of OA in women.

Another risk factor which has been identified to explain the gender difference is knee morphology (Hunter et al., 2005; Huston et al., 2000). An in-vitro study demonstrated that knee specimens from males were larger than for females with the epicondyle to epicondyle distance being 10% larger for men (Csintalan et al., 2002). Researchers investigating the muscular protection of the knee during torsion found that with age and size matched males and females, maximal knee rotation was greater in women (Wojtys et al., 2003). This would indicate an increased knee laxity in women. Knee height (heel undersurface to proximal patella) was compared with radiographic OA results and an association was identified. As knee height increased, the prevalence of radiographic OA increased, with the relationship being stronger for females (Hunter et al., 2005). Knee height may seem like a trivial factor but it allows for an estimate of the moment arm length of ligaments inserting at the knee joint, which in turn will affect the moments produced during activity.

2.2 Stair Kinematics

Climbing more than ten flights of stairs in a day has been shown to increase the risk of knee osteoarthritis (Cooper et al., 1994). Stair-climbing is a common choice of activity for research studies looking at the function of the lower limbs. It is an activity which is encountered on a daily basis by virtually everyone, making stair-climbing studies more applicable to the general population compared to activities such as squatting. Several researchers use walking studies to conduct research, but walking does not put the same magnitude of stresses on the knee. Gait is a good activity for understanding the general function of the lower limb, but when trying to understand the mechanisms of disease and injury which respect to a specific joint, it is necessary to stress that joint in a safe and structured activity. A stress on the joint, such as stair-climbing, is what is believed to challenge its integrity and lead to future problems.

It is important when collecting kinetic and kinematic data that the results be reliable and repeatable. One study has been completed which looks at the reproducibility specifically during stair-climbing. There was a significant decrease in the reproducibility of the joint angles and moments during the first step of the ascent and the last step of the descent, but the second step data was comparable between subjects (Yu et al., 1997a). In this study the subjects were walking on a level surface for the steps preceding the first stair in the ascent trials and following the last step of the descent trail. They attributed the varying patterns to motor performance as the subjects were adjusting to the difference in activity being performed (Yu et al., 1997a).

One of the first biomechanical analyses completed on stair ascent and descent was done by McFayden and Winter (1988). Gait patterns during stair-climbing are classified differently than the walking gait cycle. It is important to understand the cycle phases as this is how most of the research being reviewed will be presented. McFadyen and Winter (1988) divided the stance phase of stair ascent into weight acceptance (WA), pull-up (PU), forward continuance (FCN), while the swing phase was divided into foot clearance (FCL), and foot placement (FP). The stance phase of descent was separated into weight acceptance (WA), forward continuance (FCN), and controlled lowering (CL), and the swing phase had leg pull-through (LP), and foot placement (FP) (McFadyen and Winter, 1988). Most of the progression made during ascent and descent occurred during ‘pull-up’ and ‘controlled-lowering’ respectively (McFadyen and Winter, 1988). This makes it apparent that the joint endures the most movement while being supported by only one leg, which creates more stress on the joint.

There have been numerous studies completed on age-related changes in gait parameters, most of which involve the walking gait cycle and use healthy elderly subjects. One study that used subjects over the age of seventy-five, looked at a variety of observed performance measures (Hamel and Cavanagh, 2004). Eight of the thirty-two subjects tested exhibited unstable characteristics during stair descent but not stair ascent (Hamel and Cavanagh, 2004). This is a logical response as there tends to be a greater risk of falling during stair descent. There could also be a strength component which will be discussed later. In this age group, there is also a significant psychological component when negotiating stairs. The participants in this study with decreased confidence used greater precautions when ascending and descending the stairs, by using the handrail more frequently or moving slower (Hamel and Cavanagh, 2004). Owings et al. (2004) found that step width variability increased by 20% in the elderly subjects, another indication of decreased confidence. An increase in the variability of each step could be a sign that the subject is unsure of their footing or feel unstable.

A kinematic study that required participants to step onto a single step determined that the lateral pelvic displacement was significantly larger when compared to level walking (Collen et al., 2005). It was also found that when alternating the leading foot, lateral pelvic displacement was asymmetrical for the left and right sides (Collen et al., 2005). This is contrary to level walking which has been determined to have symmetrical pelvic sway for normal gait (Collen et al., 2005). This finding relates well to the above step width variability. As step width increases so would the lateral movement of the pelvis. This also coincides with the findings from a study involving body mass transfer during stair ascent and descent. It was determined that the divergence between center of mass and center of pressure was greater during stair-climbing when compared to level walking (Zachazewski et al., 1993). Divergence signifies greater instability meaning that the balance control would be more demanding for the body as a whole as well as proprioception specifically at the joint. In order to maintain balance, the body compensates by increasing the step width. When looking specifically at the stability requirements between ascent and descent, it appears that ascent demands less balance control (Zachazewski et al., 1993). This is shown by the double support accounting for 34% and single support at 31% compared to descent phases being 29% and 39% respectively (Zachazewski et al., 1993). The increased forward momentum would play a large role in this.

Another reported trend is that there is a larger varus angle in the knee during stair ascent, compared with descent or level walking (Yu et al., 1997b). This finding has been associated with the increased knee flexion angles seen while climbing stairs (Grood et al., 1988). It may also be linked to the increased step width and lateral pelvic displacement mentioned above.

The dimensions of the staircase being used can play a role in the outcome of the data. The dimensions of three different staircases were examined for changes in the kinematic profiles of women (Livingston et al., 1991). The steepest staircase saw the fastest cadence and least amount of time spent in double support (19-32%), which demonstrates the highest instability (Livingston et al., 1991). The middle range staircase, which is most similar to the one being used in this study, consistently had greater minimum and maximum flexion angles compared to the other two (Livingston et al., 1991).

There have been height trends associated with some of the differences seen in stair kinematics, which can translate to gender differences as the majority of women tend to be shorter than their male counterparts. Livingston *et al*., (1991) reported that shorter women had the tendency to walk with a higher cadence than taller women. Another velocity related difference was seen with elderly subjects, who took longer to reach heel contact (Collen et al., 2005). Speed can be a difficult factor to interpret as it depends on height, strength and coordination of the individual. Another reported difference attributed to height was that during both ascent and descent, women had greater peak flexion angles (Hughes et al., 2000). This is logical as people with shorter legs would have to bend their legs to a greater extent in order to clear the step.

2.3 Stair Kinetics

The kinetics of the knee displays the most change when comparing stair-climbing to level walking. The loading and moments are significantly increased during many phases of stair-climbing and the electromyography activation patterns are altered.

2.3.1 Joint Loading

It is difficult to compare joint loading between subjects as so many factors play a role in how the joint reacts to movement. Differences in gait patterns and muscle activation vary greatly between subjects, making joint loading very individualized (Taylor et al., 2004).

Knee joint loading is the most significant factor in the search for specific mechanisms causing knee osteoarthritis. Excessive loading at a joint has been demonstrated to cause injury, which later puts that joint at an increased risk of developing arthritis (Luepongsak et al., 2002). A study by Costigan *et al.* (2002) found that the net peak forces between stair ascent and level walking were not significantly different. However, the peak force was occurring at a different stage in the gait cycle. It occurs during the stance phase when the knee is at 20º flexion in level walking but occurs at 60º flexion during stair ascent (Costigan et al., 2002). With the knee in a higher flexion angle, the contact area decreases which concentrates the stress at the joint into a smaller area. Taylor, *et al.*, (2004) stated that when the knee is flexed more than 15º, both the shear and contact loads at the knee rapidly increase. This potentially leads to more wear of the joint cartilage.

Patellofemoral contact forces were also demonstrated to increase during stair-climbing compared to level walking (Costigan et al., 2002). This increased force would lead to additional pain for those with osteoarthritis. Going down the stairs has been reported as being among the most painful activities for people with knee and hip OA (Luepongsak et al., 2002). The reports of pain can be justified by biomechanical evidence.

Luepongsak, *et al.* (2002) have evidence to show that walking and stair descent are the two most common activities that elicit the greatest forces and torques at the knee, compared to standing, a chair rise and bending. Contact forces at the knee were reported by one group to range from 2.97 to 3.33 body weight (BW) for level walking and between 5.09 and 5.88 BW for stair ascent (Taylor et al., 2004). This differs from the findings from Costigan (2002). However, Costigan *et al.* (2002) measured the net force which takes into account pull from muscles, while Taylor *et al.* (2004) examined the contact force in the knee joint. When comparing these forces to those at the hip, there is a 33% higher peak force in the knee (Taylor et al., 2004). Costigan, *et al.*, (2002) found the compressive forces at the knee to average 3 BW but did reach as high as 6 times BW for stair-climbing. This is supported by the finding that the compressive forces at the knee were much higher than the shear force vector at the knee or compressive force vectors at other joints (Luepongsak et al., 2002). This is a clear sign as to why knee OA occurs more frequently than the hip. Looking at the ground reaction force, it appears that the vertical component was higher for the descent than the ascent which supports the above statements (McFadyen and Winter, 1988). In an elderly population, the ground reaction force profiles during stair descent demonstrated that the vertical loading rate in elderly women was significantly increased, maybe due to a lack of control (Hamel et al., 2005). These forces can be broken down further into different components. Shear forces were reported to be more than double when compared to those of walking (Taylor et al., 2004). Anteroposterior (AP) peak shear forces were seen to be 0.6 BW during walking and 1.3 BW during stair-climbing (Taylor et al., 2004). Costigan, *et al.*, (2002) did report the AP shear forces to be higher than those during walking but found they were slightly less than 1 BW which is lower than the value stated above. Findings from a squatting study, not consistent with the above findings, demonstrated that low anterior shear forces were observed between 0-60º knee flexion (Escamilla, 2001). However, this same study also found that tibiofemoral shear forces and patellofemoral and tibiofemoral compressive forces all increased during flexion and decreased during extension, with the peak values being reached at the highest angle achieved (Escamilla, 2001). Much of the force at higher angles is attributed to muscles. As knee flexion increases, the pull of the quadriceps increases as well. This will increase the forward and upward pull on the tibia (Costigan et al., 2002). The vertical component of this pull causes a greater axial force along the tibia, which increased the maximum force from 1 BW to as high as 4 -5 BW (Costigan et al., 2002). This would cause greater compressive forces at the knee joint. The activation of the quadriceps will also increase the patellofemoral force. The patellar force in this study was on average 3 times BW when the knee was flexed above 60º (Costigan et al., 2002). The above research supports wear and tear of the joint when stair-climbing which could lead to the onset of OA.

2.3.2 Moments Acting on the Knee

Moments acting on the knee during movement can be challenging to interpret. When comparing the knee moments during walking to those occurring while ascending stairs, the largest moments appeared at 20º (0° being full extension) during walking and at 60º during stair-climbing (Costigan et al., 2002). Similar findings were reported by Andriacchi, *et al.*, (1980) with the highest moment occurring at 50º flexion. As was noted above, this is the same angle at which the patellar force was highest showing us a trend. Luepongsak, *et al.*, (2002) reported seeing the highest torques during walking and stair descent, when compared with other activities of daily living. The contact forces in the joints are directly proportional to the net reaction moments occurring about the joint (Andriacchi et al., 1980). Because forces at the joint are such a major concern in the development of the disease, it is necessary to also incorporate the moments to fully understand what is happening at the knee.

The highest torques acting across the knee joint were the flexion and adduction reaction torques (Luepongsak et al., 2002). There is a great deal documented research on knee flexor and extensor moments, which are controlled by the hamstrings and quadriceps. This is a critical part of the research required for this study. Flexion moments at the knee during descent have been documented as being three times greater than the moments seen during level-walking (Andriacchi et al., 1980). The highest flexion moments during ascent were reported as being between 0.69 Nm/kg and 1.50 Nm/kg (Costigan et al., 2002). The flexion moment during ascent increased rapidly, to roughly 1 Nm/kg, during the first 20% of the stance cycle, while the swing leg was preparing to lift up to the next step (Costigan et al., 2002). A large flexion moment was also shown by Andriacchi, *et al.*, (1980) in the first third of the stance phase but was not as rapid as stated above. The increase was sharp and short due to the leg only flexing for a short time before the knee has to extend as the body progresses upward. Keep in mind, this is not the peak moment seen during the cycle. This was a secondary spike in moment, next to the peak experienced at 60º. The ascent peak moment is still not the highest seen during stair-climbing. Andriacchi, *et al.*, (1980) found that the flexion moment was largest during descent, but only with respect to specific stairs. The flexion-extension moment ranged from -20 to +20 Nm during the descent from step two to the floor, but the moment ranged from -40 to +120 Nm from step one to step three (Andriacchi et al., 1980). This considerable difference is due to the subject bringing both legs to the same level when reaching the floor compared to bring the leg to a lower level in the other scenario.

The research involving the knee extensor moments is not as consistent as the findings for flexion moments. But there have been more studies documenting the extension moment between different populations. Hughes, *et al.*, (2000) found that women generated a greater maximum knee extension moment than men during stair ascent. In studies looking at elderly populations, one found the extension moment in young females was much greater than that of the elderly females during stair ascent (Okita and Cavanagh, 2001). However, during stair descent, the elderly women had a higher extensor moment, which is believed to be an attempt to control the lowering of the body (Hamel et al., 2005). Another study found that during the stance phase of stair descent, 80 year old men had a higher knee extensor moment compared to the men in their 60's, although the elderly group did have much more consistent net moments in the knee (Hood and Nicol, 2005). The difference in reported findings could be a discrepancy in data analysis or it could signify that there is a significant difference in how the joint functions between men and women. When comparing descent to ascent, the extension moment during descent was double the extension moment during ascent (Andriacchi et al., 1980). Therefore, stair descent may be more critical in looking at wear and tear on the knee joint.

When looking at other moments affecting the knee, the adduction moment and internal/external moments were similar between stair-climbing and level walking (Andriacchi et al., 1980); (Costigan et al., 2002). Although, when the walking and stair descent were compared with standing, it was found that adduction moments increased eight to eleven fold (Luepongsak et al., 2002). Yu, *et al.*, (1997b) also determined that there was no significant increase in the valgus/varus (knock kneed/bow legged) knee moment at the knee between walking and stair-climbing, which is the same as the adduction/abduction moment mentioned above. Therefore, the findings are consistent between studies for the aforementioned moments. However, it was revealed that the magnitude of the valgus moment had a significant relationship with the ground reaction force in the mediolateral direction (Yu et al., 1997b). This is a logical finding as the valgus/varus action of the knee is in the mediolateral plane. Additionally, there has been no evidence to show a gender difference in peak knee adduction moments (Luepongsak et al., 2002).

The hip is not the focus of this study but it does interact with the mechanics of the knee; therefore it does warrant some attention. The hip flexion/extension moments were also reported to be highest during descent, being about one and a half times larger compared to walking (Andriacchi et al., 1980). No significant difference was seen between walking and ascending, but for both ascent and descent, the maximum hip flexion moments were generated between 30 and 40 degrees flexion (Andriacchi et al., 1980).

2.3.3 Muscle Activity and Strength

Muscle activity plays the largest role in dictating how the joint moves and the force it sustains. Andriacchi, *et al.*, (1984) stated that muscles are dependent on the direction and magnitude of the force, joint position and the combination of external moments. There have been numerous studies looking at the changes in electromyography (EMG) activation patterns produced by the muscle during different activities as well as with different populations. The myoelectric activity of the flexor and extensor muscles is dependant on the angle of the knee during a constant load (Andriacchi et al., 1984). As the knee flexes, the quadriceps muscle will lengthen and the hamstring lever arm will shorten, which will alter the efficiency of the muscle during movement.

Muscle efficiency is typically quantified using a co-contraction index. The co-contraction index that will be relevant is the one between the quadriceps (agonists) and the hamstring muscle group (antagonists). The linear envelopes of the agonist and antagonist raw EMG are integrated and summed to produce the total EMG (Winter, 2005). The Co-contraction index (CI) is the percentage difference between the total EMG and the antagonist linear enveloped EMG (Kellis, 1998). When the co-contraction index is high, the joint is considered to be inefficient. The co-contraction index was used in one study to compare the muscle function of healthy versus people with OA during stair-climbing and walking. They found that elderly with OA had 1.6-fold (BF/VL biceps femoris/vastus lateralis) ratio and 1.9-fold (BF/BF_{max}) ratio higher hamstring coactivity than the healthy adults and 2.6 and 2.8-fold increase in coactivity compared to the young healthy group (Hortobagyi et al., 2005). Clearly, the elderly diseased knees are less efficient and using more muscle activity to, most likely, reduce pain during movement or protect the knee. Both methods of representing the ratios are acceptable, but the BF/BF_{max} ratio is better to use for people with neuromuscular disorders and elderly having quadriceps weakness as BF/VL ratio could induce a misleadingly high ratio due to the weakness (Hortobagyi et al., 2005). With healthy individuals it would be more beneficial to use the BF/VL ratio as it gives a better representation of both the extensor and flexor activity on the knee.

When walking up the stairs, the body requires pulling and pushing up by the legs, which is controlled concentrically by the rectus femoris, vastus lateralis, soleus and medial gastrocnemius (McFadyen and Winter, 1988). When descending, the opposite occurs. The same muscles work against gravity to control the limb by eccentrically contracting (McFadyen and Winter, 1988). It has been well documented that stair ascent requires greater mean activity levels for the majority of muscles acting on the knee and hip (McFadyen and Winter, 1988; Lyons et al., 1983). However, there appears to be one discrepancy between these two studies. McFadyen, *et al.*, (1988) reported that the gluteus medius did not have increased activity during ascent but Lyons, *et al.*, (1983) did find increased gluteus medius activity during ascent. Both studies were looking at three modes of locomotion; however McFadyen *et. al.*, (1988) looked at normal walking while Lyons, *et al.*, (1983) looked at fast walking, which could explain the difference in findings. Other studies have investigated gluteus function. One study that looked at forward and retrograde stepping, found that gluteus maximus did not increase with retrograde stepping, which is what they predicted (Zimmermann et al., 1994). The highest gluteus maximus activity was seen during the fastest forward stepping cadence (Zimmermann et al., 1994). When looking at muscle activity with different stair inclinations, gluteus maximus, vasti, rectus femoris, and biceps femoris all increased with increasing inclination (Muller et al., 1998).

Many studies have looked at the activation levels and strength of the quadriceps muscle during activities. The knee extensors, or the quadriceps muscle group, have the greatest generation of energy during the ‘pull-up’ phase of stair ascent (McFadyen and Winter, 1988). This peak activity in the knee extensors, as well as the hip extensors, plays a large role in foot clearance (Muller et al., 1998). They act to support the leading leg. When the leg is required to raise to a higher level in order to clear the step, more muscle activation is required in the contralateral leg in order to maintain balance. The muscle activity then peaks briefly at the end when the stance leg pushes up to initiate swing (Muller et al., 1998). During descent, the quadriceps plays a significant role to control the lowering of the leg and absorb the forces at the knee during the weight acceptance phase (McFadyen and Winter, 1988). The three significant muscles that are commonly collected when looking at quad activity are the vastus lateralis, vastus medialis and rectus femoris. During the forward and retrograde stepping study, the vastus medialis appeared to have much higher EMG activity than the rectus femoris (Zimmermann et al., 1994). McFadyen, *et al.*, (1988) exhibited a similar trend with the vastus lateralis having higher muscle activity than rectus femoris during stair ascent and descent. It has been stated that this is resisted by the tensor facia latae and gastrocnemius muscle in order to protect the knee (Andriacchi et al., 1984). A study looking at gender differences at the knee during a side-step cutting task displayed greater average EMG for the quadriceps in females (vastus lateralis muscle) (Sigward and Powers, 2006). A few studies have investigated the quadriceps function in elderly. Hortobagyi, *et al.*, (2005) found that elderly subjects with OA performed activities of daily living and had significantly higher EMG activation of the quads. More of the muscle is being used due to the decrease in strength which has been reported previously. It has also been found that healthy elderly subjects exhibited an earlier onset of quadriceps activity then a young control group (Hinman et al., 2005).

It was stated above that the quadriceps muscle pulls at an angle on the tibia upward and inward.

The hamstrings are important for flexing the knee during the stance phase of the descent and the swing phase during ascent. The two hamstring muscles which are usually included in EMG collection of knee analysis studies are the biceps femoris and semitendinosis. Occasionally, the semimembranosis is collected but can be difficult to isolate due to the close proximity to the semitendinosis; therefore the two can be grouped and analyzed as one muscle as it has similar function and attachment. Retrograde stepping produced the highest activity in the semimembranosis/semitendinosis but when cadence increased in forward stepping this was the only muscle that did not increase significantly in activity (Zimmermann et al., 1994). Similar findings were reported by Colliander, *et al.*, (1989) who found that hamstring peak EMG showed no change with increasing angular velocity in men, but women did show increased activity. The findings by Zimmerman *et. al.,* (1994) could have been confounded by the men and women being analyzed as one group. Another study that looked at hamstring activity showed that during stair ascent, the rectus femoris and semimembranosis were active at the same time during the cycle but had different intensities (Lyons et al., 1983). The biceps femoris was most active during loading and just before swing was initiated and the semi was most active during mid swing (Lyons et al., 1983). Mid-swing is when the knee is flexing to clear the step up. This timing of the semitendinosis is similar to that reported by McFadyen, *et al.*, (1988). During descent, the semitendinosis was most active during the stance which is when the leg supports the body while it is being lowered (McFadyen and Winter, 1988).

As can be seen from the evidence above, the knee endures the highest forces during descent but the muscles have the highest activity during ascent. With the muscular support lacking during descent to absorb the high forces, it is evident how stair activities could lead to damage to the joint over many years.

2.4 Quadriceps Muscle Angle

As can be seen from above, the quadriceps plays a significant role in how the knee functions during activity and the capacity to which the knee absorbs compressive and shear forces. It should also be evident that female mechanics are different than those of a male which explains the increased prevalence of OA in women. One anatomical factor that has been shown to be significantly different between men and women is the quadriceps muscle angle; however this has not been connected to tibiofemoral problems. The quadriceps muscle angle, or the q-angle, is a measure of the alignment of the quadriceps femoris muscle relative to the underlying skeletal structures of the pelvis, femur and tibia (Livingston and Spaulding, 2002). It is thought that higher q-angles increase the lateral pull from the quadriceps on the patella and the tibia. The measurement of the q-angle is done by calculating the angle between a line connecting the anterior superior iliac spine (ASIS) and the center of the patella, representing the quadriceps, and another line connecting the tibial tuberosity to the center of the patella, representing the patellar tendon (Horton and Hall, 1989; Livingston, 1998; Mizuno et al., 2001).

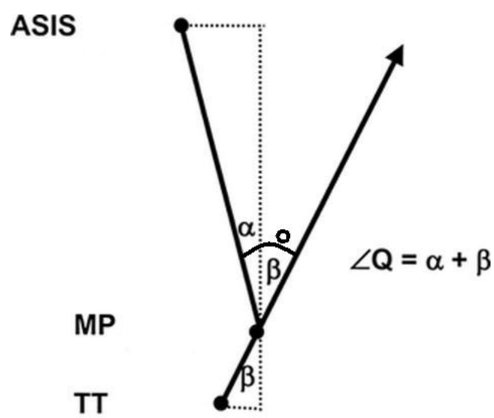


Figure 2.1: *Q-Angle and Vectors Created by the Anterior Superior Iliac Spine (ASIS), Mid-Patella (MP) and Tibial Tuberosity (TT) of a right leg. (Livingston and Spaulding, 2002).*

The position of the subject while the measurement is being taken is a critical factor as the q-angle magnitude increases and decreases with knee movement and muscle activation (Livingston and Spaulding, 1999; Lathinghouse and Trimble, 2000). As knee flexion increased up to a certain point, the q-angle would increase to as high as 25-30º, but would then decrease with further flexion (Livingston and Spaulding, 1999). This u-shaped relationship is most likely due to the hypothesized screw home mechanism of the knee. The screw home mechanism of the knee is caused by uneven articular surfaces between the medial and lateral sides of the tibia. During knee flexion, the amount of internal/external rotation differs depending on the angle and corresponding articular surface. Increased muscle activation of the quadriceps has produced smaller q-angles (Lathinghouse and Trimble, 2000), as well as increased over-all strength being associated with differences in q-angle (Bayraktar et al., 2004). Individuals with increased peak torque had a decreased magnitude in q-angle (Byl et al., 2000). It has also been shown to change with supine and standing positions, with standing exhibiting larger q-angle values (Woodland and Francis, 1992) as well as seeing a decreased q-angle after the insertion of an orthotic shoe device (Kuhn et al., 2002). One study was completed in an attempt to create a standard for measurement. The q-angle was measured using motion analysis markers with the feet placed in three different positions. It was determined that the ideal standing position was the Romberg stance (Livingston and Spaulding, 2002). The Romberg stance is when the feet are together with the medial borders (Livingston and Spaulding, 2002). In most of the studies examined here, the subjects stood weight-bearing with the knees fully extended and the quadriceps muscle relaxed, as this is when individuals are most likely to experience problems (Livingston and Spaulding, 2002; Byl et al., 2000; Lathinghouse and Trimble, 2000). One other factor important in the measurement of the q-angle is that it is not symmetrical between legs (Livingston and Mandigo, 1999); therefore it is critical that both legs are measured and reported during investigations.

The key finding in q-angle research is surrounding gender. Numerous studies determined that women statistically have a higher q-angle than men (Woodland and Francis, 1992; Byl et al., 2000; Horton and Hall, 1989). And women have typically been considered to fall in the excessive measures, which according to the American Orthopedic Association, is between 15º and 20º. Other sources claim that normal angles in men range from 8°-14° and from 11°-20° in women (Horton and Hall, 1989). Recently, values between 8° - 10° for men and up to 15° for women are considered normal and that values above this can be problematic (Livingston and Spaulding, 2002; Greene et al., 2001; Livingston, 1998; Boden et al., 1997). Lathinghouse, *et al.*, (2000) reported that an excessive q-angle may predispose women to greater lateral displacement of the patella during activity requiring high levels of quadriceps activation. It was suggested that the higher q-angles in women were due to them having wider hips, but this is not the case. In fact, one study reported that men had wider hips than women when measuring greater trochanter to greater trochanter (Horton and Hall, 1989). It has been determined that hip breadth or femur length do not account for the discrepancy in q-angle between men and women (Byl et al., 2000; Horton and Hall, 1989). However, the hip width-femur length ratio is slightly lower in men although it has not been scientifically correlated to q-angle (Horton and Hall, 1989).

Q-angle has typically been the focus of the research surrounding patellofemoral disorders. Only recently has the q-angle been associated with tibiofemoral mechanics. When the q-angle was decreased, the valgus orientation of the knee decreased as well (Mizuno et al., 2001). This could reduce the tibiofemoral contact pressures in the lateral compartment of the knee, but with the reduced valgus, the varus orientation produced could increase medial contact forces. There was also a lateral tibial translation reported with increased q-angle (Mizuno et al., 2001).

3. Study I: Repeatability of Q-Angle Measurements: Interclass and Intraclass Correlations

3.1 Methods

3.1.1 Subjects

Twenty university aged subjects (10 males, 10 females) were recruited from the University of Waterloo student and staff population (18-30 years of age). The q-angle was measured by three individuals: the primary researcher and two clinicians (a Chiropractor and Physiotherapist) to determine the inter-rater reliability. Additionally, it was measured on three separate days by the primary researcher to determine the intra-rater repeatability.

3.1.2 Equipment and Protocol

The q-angle measurement was collected using 3D Optotrak Certus™ motion capture system (Northern Digital Inc., Waterloo, ON). To test the inter-rater reliability, one of the three researchers placed IRed (infra-red) markers on the subject's skin using double sided adhesive, identifying the anterior superior iliac spine (ASIS), tibial tuberosity and the mid-patella (Figure 3.1). The mid point of the patella was determined by the intersection of a line drawn from the medial to the lateral patella and a second line from the inferior to superior patella (Lathinghouse and Trimble, 2000). The three marker positions on both legs were collected simultaneously with the subject standing in Romberg Position: medial borders of the feet touching with knees fully extended and quadriceps muscle relaxed. This static collection was repeated three times to accommodate for sway, without altering the markers. The three q-angle measurements were averaged and this value was used in the analysis. The markers were removed and this process was repeated by the other two researchers. To test the intra-rater reliability, the subjects returned on three separate days for testing, being tested by the primary researcher each day.

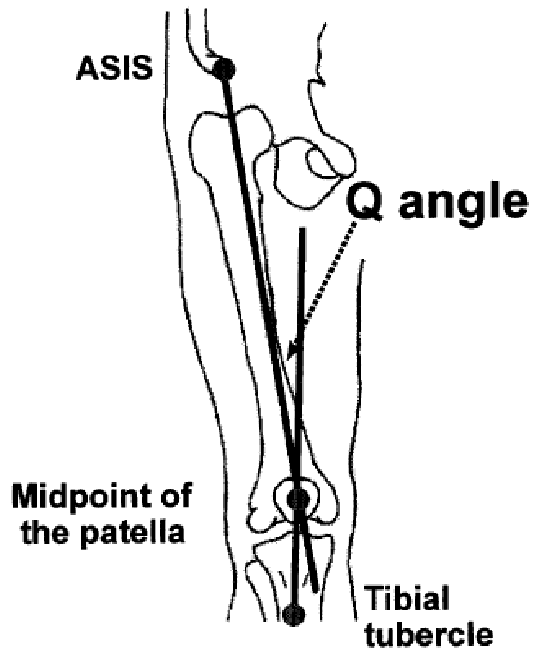


Figure 3.1*: Q-Angle and Marker Locations: Anterior Superior Iliac Spine (ASIS), Mid-Patella (MP) and Tibial Tuberosity (TT) (Livingston and Mandigo, 1999). Anterior View.*

3.1.3 Data and Statistical Analysis

The q-angle values were calculated using the X, Y, Z coordinates from the Optotrak data (accuracy of 0.1mm at 2.25m). A vector was created connecting the ASIS to the mid-patella marker and a second vector was created between the mid-patella and the tibial tuberosity. These 3D vectors were then projected onto a 2D frontal plane defined by the orientation of the pelvis and thigh segments. The 3D coordinates were first rotated about the transverse axis, aligning a line connecting the ASIS markers with the frontal plane of the global coordinate system. The coordinates were then rotated about a mediolateral axis, aligning the ASIS and mid-patella markers of each thigh with the longitudinal axis of the global coordinate system. The angle between the two vectors was the resulting q-angle (Figure 4.4). This was done using custom software. The interclass correlation coefficient was calculated to determine the accuracy of the measurement by different observers and the intraclass correlation coefficient was calculated to determine the repeatability of the q-angle measurement by the same observer (Shrout and Fleiss, 1979).

The intra-rater calculation was calculated using:

$$ICC_1 = \frac{BMS - EMS}{BMS + (k-1)\,EMS}$$

The inter-rater calculation was calculated using:

$$ICC_2 = \frac{BMS - EMS}{BMS + (k-1)\,EMS + k(JMS - EMS)/n}$$

BMS is the subject mean square, EMS is the mean square error and JMS is the mean square of the rater. k is the number of raters and n is the number of subjects.

3.2 Results

The q-angle data for the intra-class and inter-class correlations was collected successfully on 8 males and 8 females. Individual q-angle measurements are presented in Table 3.1 along with the ICC values and means $\pm$ standard deviation (SD). The intraclass correlation coefficients (ICC_1) were all above 0.80 as can be seen in Table 3.1. An ICC of greater than 0.80 is considered to be an excellent level of repeatability (Fleiss, 1986). This demonstrates a high reliability of the q-angle measurement from the primary researcher.

Two of ten participants were missing for both the male and female groups. This was due to an incorrect camera file being used during the data collection. The inter-rater reliability (ICC_2) proves to be much less reliable with ICC values ranging from 0.27-0.78 as shown in Table 3.2. None of the four interclass measurements were considered to have a good level of reliability.

***Table 3.1**: Individual Q-Angle Measurements: Mean Q-Angle Measurements and Intraclass Correlation Coefficients (ICC). Each subject had three repeated measurements taken by the same rater on different days.*

		Females (n = 8)		**Males (n = 8)**	
SUBJECT		**Right Q-Angle**	**Left Q-Angle**	**Right Q-Angle**	**Left Q-Angle**
1	Day 1	12.56	7.43	12.77	8.54
	Day 2	11.78	7.87	13.69	6.12
	Day 3	12.80	5.75	15.23	10.74
2	Day 1	20.93	21.93	16.38	8.87
	Day 2	23.40	28.29	17.52	15.27
	Day 3	19.64	22.00	20.02	16.87
3	Day 1	39.96	42.89	15.28	8.43
	Day 2	32.52	39.69	10.46	10.05
	Day 3	32.76	37.15	14.06	6.79
4	Day 1	14.56	13.48	7.47	13.88
	Day 2	15.59	14.12	13.13	10.67
	Day 3	15.90	11.85	7.91	10.45
5	Day 1	18.87	20.41	15.18	16.04
	Day 2	19.29	23.74	15.71	14.63
	Day 3	11.92	27.33	10.95	16.05
6	Day 1	12.07	11.75	14.78	21.24
	Day 2	11.23	12.53	10.40	18.26
	Day 3	12.56	13.87	15.71	14.86
7	Day 1	10.11	5.41	20.35	14.80
	Day 2	17.60	20.69	17.81	10.16
	Day 3	18.35	13.26	16.06	16.84
8	Day 1	19.03	15.92	8.29	16.46
	Day 2	17.74	14.74	11.11	16.93
	Day 3	21.43	26.22	10.27	19.09
Mean Q-Angle ± SD		18.44 ± 7.49	19.10 ± 10.39	13.77 ± 3.58	13.42 ± 4.15
ICC Intra-rater		0.95	0.95	0.80	0.82

__Table 3.2__: Individual Q-Angle Measurements: Mean Q-Angle Measurements and Interclass Correlation Coefficients (ICC). Each subject had three repeated measurements taken by a different rater. First value is rater 1, second is rater 2 and third is rater 3.

		Females (n = 8)		Males (n = 8)	
SUBJECT		**Right Q-Angle**	**Left Q-Angle**	**Right Q-Angle**	**Left Q-Angle**
1	Rater 1	11.78	7.87	15.23	10.74
	Rater 2	16.25	8.05	14.92	9.99
	Rater 3	28.96	8.76	13.46	23.64
2	Rater 1	23.40	28.29	20.02	16.87
	Rater 2	18.89	22.87	23.36	15.61
	Rater 3	23.46	16.59	21.06	27.12
3	Rater 1	32.52	39.69	14.06	6.79
	Rater 2	38.54	44.84	7.20	4.75
	Rater 3	32.75	40.58	9.74	6.82
4	Rater 1	15.59	14.12	7.91	10.45
	Rater 2	12.23	14.11	8.88	10.65
	Rater 3	8.19	27.14	9.83	14.05
5	Rater 1	19.29	23.74	10.95	16.05
	Rater 2	16.13	19.03	13.92	22.89
	Rater 3	13.01	30.70	11.41	27.97
6	Rater 1	11.23	12.53	15.71	14.86
	Rater 2	25.79	16.86	21.03	15.82
	Rater 3	24.36	24.66	15.01	20.87
7	Rater 1	17.60	20.69	16.06	16.84
	Rater 2	25.77	12.64	12.66	19.55
	Rater 3	18.14	22.60	10.76	22.35
8	Rater 1	17.74	14.74	10.27	19.09
	Rater 2	21.15	28.06	6.72	23.03
	Rater 3	21.58	28.83	6.41	27.16
Mean Q-Angle ± SD		20.60 ± 7.53	22.00 ± 10.23	13.19 ± 4.74	16.83 ± 6.73
ICC Inter-rater		0.27	0.71	0.78	0.62

3.3 Discussion

Overall, ICC_1 values showed a high degree of intra-rater reliability (0.80-0.95); however ICC_2 demonstrated a low degree of reliability for the inter-rater reliability (0.27-0.78). ICC values from other studies had intra-class correlations ranging from 0.89-0.98 (Livingston and Spaulding, 2002; Lathinghouse and Trimble, 2000; Horton and Hall, 1989; Shultz et al., 2006; Tomsich et al., 1996) and inter-class correlations ranging from

0.67-0.79 (Livingston and Mandigo, 1999; Horton and Hall, 1989; Shultz et al., 2006; Tomsich et al., 1996). The intra-rater values achieved here are similar to the above studies; however, the inter-rater correlations are far below the values above. One well-cited study which looked at the reliability of 6 anatomical measurements had ICC values of 0.63 for intra-rater reliability and 0.23 for inter-rater reliability for the q-angle (Tomsich et al., 1996). Those inter-class correlations correspond much better to the values exhibited in this study.

Low inter-class correlations (ICC_2) make it difficult to set standards or compare different study results. The q-angle has been a clinical measure for more than 20 years and there is still not a clear understanding of what is considered to be an excessive measure or normal (Lathinghouse and Trimble, 2000; Livingston and Spaulding, 2002). It also makes ICC_2 unreliable as a clinical tool for accurately diagnosing and treating patients. The high intra-class correlations still warrant the use of this measure, such as in situations where the measure is being used in a functional study and there is only 1 rater. If the rater typically measures a higher than average q-angle, it will be higher for everyone; therefore, the data set will still be comparable.

One likely explanation for the low inter-class correlation (ICC_2) values is different strategies used to locate the three landmarks on each leg. Two of the four participants that were unusable appeared to have incorrect marker placement yielding a negative value. This occurs when either the tibial tuberosity marker is placed too far medially or the mid-patella marker is placed too far laterally. The two clinicians involved in the study were a physiotherapist and a chiropractor. Physiotherapists typically measure the q-angle supine but it is becoming more common for clinicians to measure in standing position as it is recognized as a better representation of quadriceps function. Clinicians are also used to measuring q-angle using a goniometer rather than the motion capture used in this circumstance. This may explain some of the difference between raters. Also, there was a discrepancy in how long the clinicians had been practicing. One has been a clinician for more than 20 years while the other had been practicing less than one year. Methodology in how they were taught could differ as well as the familiarity with land-marking this area in particular.

The mid-patella can be a challenging place to landmark due to the surrounding tissue. An issue related to this was that the participants felt it was challenging to maintain relaxed quadriceps during the Romberg stance. They were activating the muscle on and off to maintain balance and reduce sway which alters the position of the patella. It has been demonstrated that the q-angle can be altered by isometric quadriceps activation (Lathinghouse and Trimble, 2000). This could still affect ICC_2 more so than ICC_1, as a particular rater could be more diligent in assuring the subject has a relaxed quadriceps during marker placement.

To summarize what was stated above, land-marking appears to be the major limitation in this study. One study which induced error into the measurement by displacing the markers by 1-5mm found that this altered the q-angle by 0.12° to 5.18°, when locating the bony landmarks (France and Nester, 2001). If the markers are put on the skin while the quadriceps are activated or there is significant movement of the subject between marker placement and collection, the data could be altered.

3.4 Conclusion

It was determined that the intra-rater correlation values had an excellent level of repeatability; however the inter-rater values were not very reliable. This was explained by improper marker placement. The overall values determined in this study were comparable to values in the literature, demonstrating the methods and data analysis used in this study to be good representation of the q-angle measurement. Using q-angle measurements from different raters is not recommended when evaluating its relationship with other variables. However, the use of the q-angle measure is reliable with a single rater.

4. Study 2: The Influence of Q-Angle and Gender on the Stair-Climbing Kinetics and Kinematics of the Knee

4.1 Methods

4.1.1 Subjects

Ten male and ten female subjects were recruited from the University of Waterloo student and staff population (18-30 years of age). Subjects were excluded if (1) they experienced pain in the hip, knee, or ankle joint more than one day per month, (2) they had undergone surgery to the hip, knee or ankle joint involving ligament, meniscus or bone, or (3) they had surgery which has damaged hip, knee or ankle ligaments, meniscus or bone. Each subject had their static q-angle measured, and then completed 20 stair-climbing trials (10 ascent and 10 descent), with 5 left foot leads and 5 right foot leads being randomized. Age, weight and height were also recorded then. All subjects completed the trials barefoot to help standardize the protocol. EMG, kinematics and kinetics were recorded first, then quadriceps and hamstring muscle activation patterns, net knee joint moments and occurrence of peak moments were calculated to determine different strategies of limb control used during stair-climbing between genders.

4.1.2 Equipment and Protocol

The q-angle was measured using 3D Optotrak Certus™ motion analysis system (Northern Digital Inc., Waterloo, ON). Each subject stood in Romberg position: the medial borders of the feet touching with legs fully extended and the quadriceps muscle completely relaxed. Markers were placed on the subject's skin, identifying the anterior superior iliac spine (ASIS), tibial tuberosity and the mid-patella. The mid point of the patella was determined by the intersection of a line drawn from the medial to the lateral patella and a second line from the inferior to superior patella (Lathinghouse and Trimble, 2000). The three marker positions on both legs were collected simultaneously, three separate times. The patella and tibial tuberosity markers were removed following the collection of the three q-angle recordings. The ASIS marker remained as this was used for the kinematic data.

Electromyography (EMG) Meditrace™ silver-silver chloride (Ag-AgCl) electrodes were placed on the subject by the primary researcher. Three muscles were collected during the study: the vastus lateralis (VL), vastus medialis (VM), and biceps femoris (BF). The vastus lateralis and biceps femoris were chosen as they have been shown to have higher activation patterns during stair-climbing and are anatomically larger muscles (Lyons et al., 1983; Zimmermann et al., 1994). Any hair covering the electrode area was removed using a single use disposable razor. The electrode area was cleaned with rubbing alcohol to remove any dirt or dead skin cells. The researcher tested to assure there was a clean signal by using manual resistance to elicit a sub-maximal contraction from each muscle. The subject was asked to sit on a stool and a trial was collected with the muscles relaxed (i.e. quiet). There were 3 maximum voluntary exertion (MVE) trials collected from each of the 3 muscles. The quadriceps (VL/VM) maximum voluntary exertions (MVE) were recorded with the subject seated, hips at 90º with the leg positioned at a 60º relative knee angle (Figure 4.1) (Perotto et al., 2005). Participants were asked to extend their knee against manual resistance from the primary researcher. The biceps femoris MVE was collected with the subject lying prone with the knee at an angle of 60° (Perotto et al., 2005). The subject was instructed to elicit a flexion contraction against manual resistance placed on the lower calf. For all MVE's, subjects were asked to ramp up to a full isometric contraction in which they were contracting as hard as they could for five seconds. Each series was completed three times, with a 2 minute rest period between each set. The EMG data was collected at 2048Hz with a 12-bit $\pm$ 5V A/D system. A 16-channel differential amplifier (Bortec™, Calgary, AB) with a 10-1000Hz bandpass filter and $\pm$ 4V peak to peak output was used.

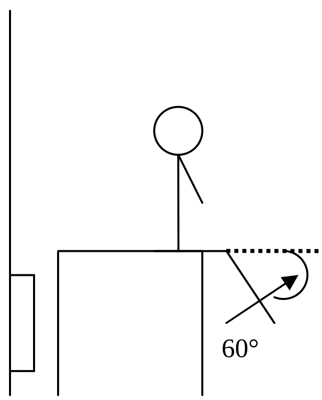


Figure 4.1*: Experimental Setup for Maximum Voluntary Exertion (MVE) Trial for Quadriceps. (0° = leg in full extension)*

The kinematic and kinetic data was collected using a 3D Optotrak Certus™ motion capture system (Northern Digital Inc., Waterloo, ON) and a 6-channel force plate (Model: OR6-7-2000, AMTI, Watertown, MA) built into the second step of a four step staircase. The steps of the staircase have a rise and run of 20cm and 30cm respectively, which were chosen according to the University of Waterloo building codes (Figure 4.2). The force plate data was collected at the same sampling rate as the EMG, 2048Hz, due to both signals being collected through the Optotrak Data Acquisition Unit (ODAU) which synchronizes these signals with the motion data.

Force Plate

Figure 4.2*: Staircase used for the ascent and descent trials. Rise = 20cm. Run = 30cm.*

Thirty-six Optotrak™ markers were then placed on the subject by the primary researcher. Sixteen of the markers were used as tracking markers which were placed on four rigid plates secured on the mid section of the thigh and shank of each leg using neoprene wraps. The remaining twenty markers were placed on the bony prominences of the right and left lower limbs: ASIS, greater trochanter, medial and lateral femoral condyles, medial and lateral tibial plateau, medial and lateral malleoli, lateral base of the heel and the distal fifth metatarsal. These markers were used for the static calibration trial which determined the location and orientation of the tracking plates with respect to the joint markers during the stair-climbing trials. The motion data was collected at a sampling rate of 64Hz. Each subject was asked to go up and down the stairs a few times until they felt comfortable with the protocol. Subjects began with an ascent trial. Ten ascent and 10 descent trials were completed. Left and right foot leads for each trial (5 left foot lead and 5 right foot lead) were randomized. Figure 4.3 shows the experimental setup.

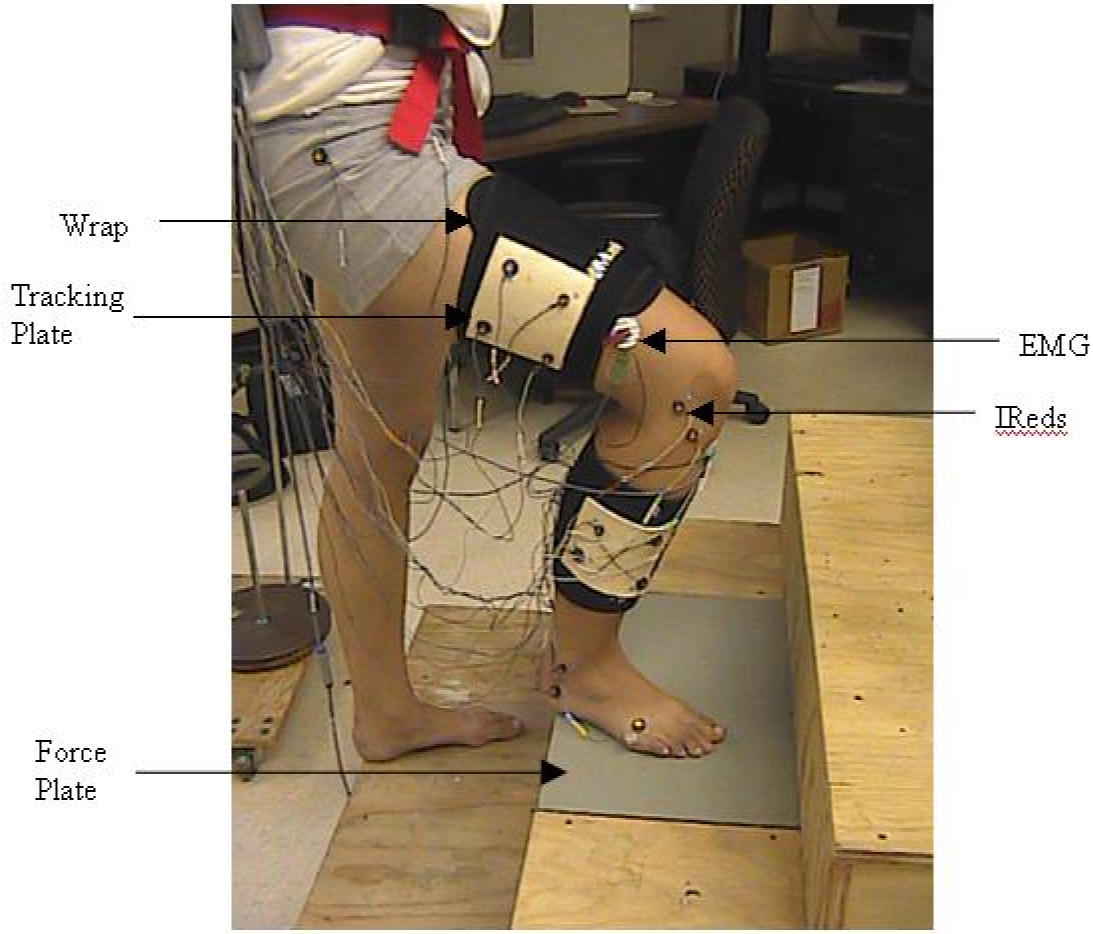


Figure 4.3: Experimental setup for the data collection. The force plate is embedded into the second step. The tracking plates, both on the thigh and shank were secured to the wraps with Velcro. The EMG was located under the neoprene wraps; one of the two electrodes was exposed. The IReds (motion markers) can be seen on the knee joint, foot, hip and tracking plates.

4.1.3 Data Analysis

The q-angle values were calculated using the X, Y, Z coordinates from the Optotrak data. A vector was created connecting the ASIS to the mid-patella marker and a second vector was created between the mid-patella and the tibial tuberosity. These 3D vectors were then projected onto a 2D frontal plane defined by the orientation of the pelvis and thigh segments. The 3D coordinates were first rotated about the transverse axis, aligning a line connecting the ASIS markers with the frontal plane of the global coordinate system. The coordinates were then rotated about a mediolateral axis, aligning the ASIS and mid-patella markers of each thigh with the longitudinal axis of the global coordinate system. The angle between the two vectors was the resulting q-angle (Figure 4.4). This was done using custom software. The three angles calculated were averaged and this was the value used in analysis. The q-angle measurements collected for each leg were grouped by excessive (high) and normal (low) values. Excessive values were considered to be >15° for women and men. Angles greater than $\geq$15-20° are considered to be excessive, therefore the low end of this range was chosen to ensure more evenly distributed high and low groups (Horton and Hall, 1989; Livingston, 1998; Greene et al., 2001; Byl et al., 2000).

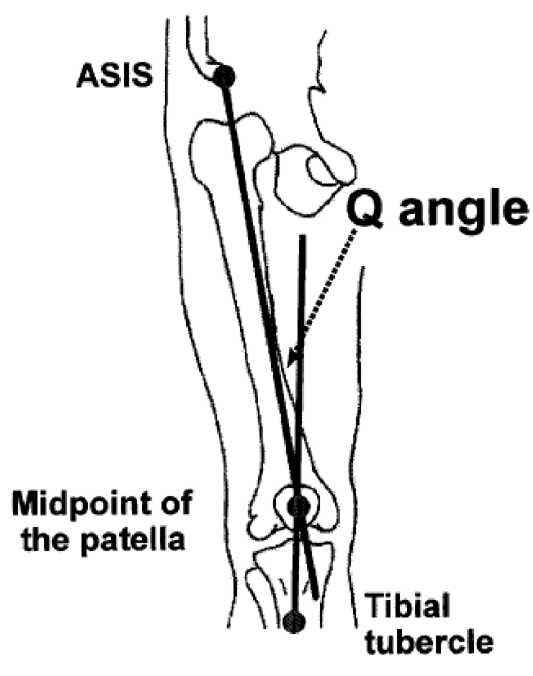


Figure 4.4: *Q-Angle and Marker Locations: Anterior Superior Iliac Spine (ASIS), Mid-Patella (MP) and Tibial Tuberosity (TT) (Livingston and Mandigo, 1999).*

The kinematic data was low-pass filtered using a 6Hz (Winter, 2005) dual pass 4th order Butterworth filter. Using a three-axis Euler angle system, 3D relative joint angles were determined for the knee. The 3D internal moments occurring about the knee and ankle were calculated from the kinetic data. The filtering and kinematic and kinetic processing were completed using Visual 3D™ software (C-Motion, Maryland, USA). Moments were normalized to body weight. The kinematic and kinetic data were then rubber banded to toe-on and toe-off of the force plate, ending up with a measure of 100% movement while in contact with the force plate. The toe-on/toe-off threshold was determined by calculating the mean and standard deviation of the first full second of the force plate recording, then looking for the point at which the signal exceeded that mean by 2 standard deviations. This data was ensemble averaged for each subject to combine the data from the 5 right ascent, 5 right descent, 5 left ascent and 5 left descent trials. Finally, the peak moment and occurrence of peak moment were calculated and used in the statistical analysis.

All EMG data was full wave rectified and low-pass filtered at 3Hz using a 2nd order low-pass Butterworth filter (McFadyen and Winter, 1988). The rest trial was averaged and this mean was subtracted from the MVE and stair-climbing trials. The stair-climbing trials were normalized to the peak values from the MVE of the respective muscle. This resulted in a measure that was normalized to a percent of the maximum exertion. The data was then rubber banded to toe on, toe off the force plate; same method used for the kinetic and kinematic data.   The 5 right ascent, 5 right descent, 5 left ascent and 5 left descent trial were ensemble averaged. The peak EMG, occurrence of peak

EMG and average EMG were calculated for all three muscles in each leg. Only the leg in contact with the force plate was analyzed for any one trial.

4.1.4 Statistical Analysis

The q-angles were grouped by gender and by excessive and normal q-angle ($\geq 15°$ = excessive, $<15°$ = normal). The EMG and kinetic variables were run through a two-way General Linear Model (GLM) with q-angle and gender as factors. A P-value of 0.05 was used. The right and left legs during the ascent/descent were combined and treated as independent samples. The variables analyzed included peak EMG, occurrence of peak EMG and average EMG for the three muscles and peak and occurrence of peak moments for the knee extension and abduction. Joint angles were not analyzed as they are dependant on height and not controlled for.

4.2 Results

Ten male and 10 female subjects were collected. The females had a mean age of 24.8 years, mass of 67.98kg and height of 1.64m. Males had a mean age of 24.7 years, mass of 81.40kg and height of 1.79m (Table 4.1).

Table 4.1: *Descriptive Statistics for Subjects in Study II.*

	Females (n = 10)	**Males (n = 10)**
Age (years) ± SD	24.80 ± 3.12	24.70 ± 2.75
Mass (kg) ± SD	67.98 ± 14.15	81.40 ± 12.57
Height (m) ± SD	1.64 ± 0.16	1.79 ± 0.05

4.2.1 Q-Angle

The average q-angle for males was lower compared to females and the measures between legs were asymmetric (Table 4.2), although not tested statistically. In the males, there were 5 excessive q-angle measures for the right leg; subjects 6, 7, 8, 9, 10; and 2 excessive measures for the left leg; subjects 6, 10. Values were not rounded up to 15°. For the females, there were 6 excessive measures for the right leg; subjects 3-7 and 10; and 5 excessive measures for the left leg; subjects 2, 3, 4, 7, 9.

***Table 4.2**: Q-Angle Values and Means for Male and Female Left and Right Leg*

Males	Right Q-angle	Left Q-angle		Females	Right Q-angle	Left Q-angle
1	9.84	7.51		1	12.80	13.33
2	10.46	10.05		2	10.81	18.21
3	13.13	10.45		3	19.64	22.00
4	8.73	11.93		4	32.76	22.68
5	11.01	14.24		5	17.36	13.15
6	15.78	25.70		6	15.90	11.85
7	15.71	14.63		7	18.87	20.41
8	20.90	10.97		8	12.56	13.87
9	20.35	10.53		9	12.17	17.83
10	19.39	15.13		10	17.74	14.74
Mean ± SD	14.53 ± 4.57	13.11 ± 5.02			17.06 ± 6.32	16.81 ± 3.95

4.2.2 Stair Ascent

There were 13 dependent variables tested: 3 average EMG, 3 occurrence of peak EMG, 3 peak EMG, 2 peak moments and 2 occurrence of peak moments. Due to the left and right limbs being treated as independent samples, there were a total of 40 samples (limbs) for stair ascent. Table 4.3 displays the mean values for the 13 dependent variables and the associated standard deviations. They have been divided into gender and high/low q-angle groups.

Table 4.3*: Means and Standard Deviations (SD) for the Dependant Variables Analyzed for the Stair Ascent Trials. Average EMG (AEMG) and Peak EMG are in units of %MVE. Occurrence of peak EMG and Moments are expressed as 0-100% stance. Peak Knee Moments are expressed in Nm/kg. VL = Vastus Lateralis. VM = Vastus Medialis. BF = Biceps Femoris. Highlighted boxes were statistically significant (p<0.05). No interaction effects were present.*

	Gender (n=40)		**Q-Angle (n=40)**	
	Females (n=20)	**Males (n=20)**	**High (n=18)**	**Low (n=22)**
VL AEMG	57.53 ± 31.60	29.95 ± 15.65	36.87 ± 25.55	47.44 ± 30.33
VM AEMG	56.43 ± 30.74	30.79 ± 17.99	40.09 ± 23.06	45.50 ± 30.65
BF AEMG	5.65 ± 2.15	5.83 ± 3.01	5.17 ± 2.78	6.04 ± 2.48
VL Occurrence of Peak EMG	21.91 ± 2.23	23.57 ± 3.37	23.98 ± 2.79	22.07 ± 2.86
VM Occurrence of Peak EMG	22.91 ± 9.35	22.48 ± 2.75	25.02 ± 10.82	21.44 ± 2.59
BF Occurrence of Peak EMG	61.193 ± 28.39	62.97 ± 24.44	62.98 ± 28.48	61.60 ± 25.40
VL Peak EMG	156.50 ± 86.40	81.78 ± 42.86	110.48 ± 69.18	123.80 ± 82.16
VM Peak EMG	153.85 ± 89.09	91.92 ± 57.85	120.27 ± 76.95	124.30 ± 83.82
BF Peak EMG	12.38 ± 4.28	14.06 ± 6.72	11.70 ± 4.30	14.05 ± 6.14
Peak Knee Extension Moment	0.72 ± 0.14	0.69 ± 0.15	0.66 ± 0.13	0.73 ± 0.15
Occurrence of Peak Knee Extension Moment	26.40 ± 1.82	25.95 ± 2.62	26.50 ± 2.85	26.00 ± 1.88
Peak Abduction Knee Moment	0.17 ± 0.10	0.23 ± 0.09	0.18 ± 0.11	0.21 ± 0.09
Occurrence of Peak Abduction Moment	49.74 ± 23.09	39.95 ± 21.87	43.23 ± 21.57	45.46 ± 23.66

Significant differences were found for gender in VL average EMG (F = 12.12, p = 0.0013), VM average EMG (F = 9.94, p = 0.0033), VL peak EMG (F = 11.69, p = 0.0016), and VM peak EMG (F = 6.53, p = 0.0149).

There was a significant difference found for q-angle in biceps femoris (BF) average EMG (F = 5.13, p = 0.0297) and knee adduction occurrence of peak moment (F = 4.50, p = 0.0410). The 'p' and 'F' values for the entire data set can be seen in Appendix 7. Of the four gender variables with a significant difference, females had higher means than the males for all of them. The means of the four significant gender variables for ascent can be seen in Figure 4.5.

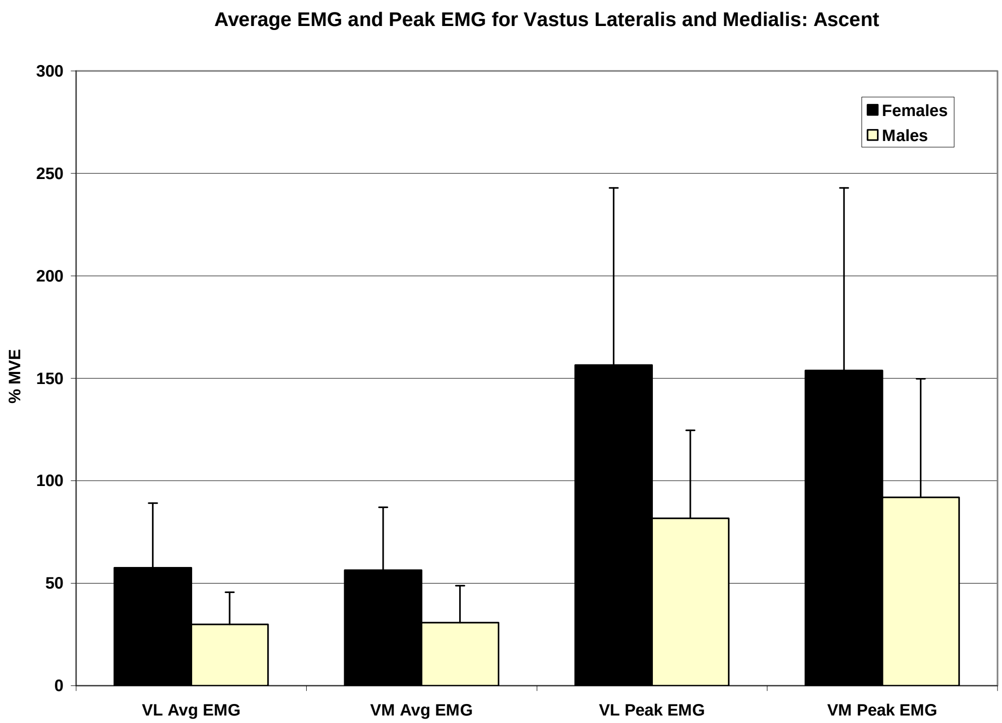


Figure 4.5*: The mean values for peak EMG and average EMG for the Vastus Lateralis (VL) and Vastus Medialis (VM). The error bars represent the '+' Standard Deviation (SD) within each variable. They are represented in terms of '% Maximum Voluntary Exertion' (MVE).*

4.2.3 Stair Descent

Table 4.4 displays the mean values for the 13 dependant variables and the associated standard deviations. They have been divided into gender and high/low q-angle categories.

***Table 4.4**: Means and Standard Deviations (SD) for the Dependant Variables Analyzed for the Stair Descent Trials. Average EMG (AEMG) and Peak EMG are in units of %MVE. Occurrence of peak EMG and Moments are expressed as 0-100% stance. Peak Knee Moments are expressed in Nm/kg. VL = Vastus Lateralis. VM = Vastus Medialis. BF = Biceps Femoris. Highlighted boxes were statistically significant (p<0.05). No interaction effects were present.*

	Gender (n=40)		**Q-angle (n=40)**	
	Females (n=20)	**Males (n=20)**	**High (n=18)**	**Low (n=22)**
VL AEMG	41.28 ± 21.14	21.08 ± 12.82	30.05 ± 22.67	31.79 ± 18.97
VM AEMG	39.57 ± 20.10	20.08 ± 11.41	28.51 ± 20.90	30.52 ± 18.14
BF AEMG	4.10 ± 1.61	3.21 ± 2.12	3.47 ± 2.02	3.76 ± 1.89
VL Occurrence of Peak EMG	80.80 ± 1.41	64.41 ± 19.12	73.86 ± 15.75	71.93 ± 16.01
VM Occurrence of Peak EMG	76.20 ± 9.85	74.71 ± 11.62	76.07 ± 13.68	76.74 ± 8.67
BF Occurrence of Peak EMG	84.82 ± 15.02	58.16 ± 33.08	86.70 ± 14.50	63.31 ± 31.29
VL Peak EMG	94.47 ± 41.52	44.99 ± 30.54	70.20 ± 46.19	69.48 ± 43.46
VM Peak EMG	86.50 ± 37.71	51.31 ± 44.44	65.47 ± 42.82	70.76 ± 45.97
BF Peak EMG	13.90 ± 8.18	9.34 ± 5.13	12.48 ± 9.95	11.16 ± 5.21
Peak Knee Extension Moment	0.73 ± 0.09	0.64 ± 0.16	0.65 ± 0.15	0.71 ± 0.13
Occurrence of Peak Knee Extension Moment	76.70 ± 12.54	73.80 ± 20.16	81.14 ± 3.06	72.08 ± 19.93
Peak Abduction Knee Moment	0.23 ± 0.10	0.32 ± 0.12	0.26 ± 0.11	0.29 ± 0.13
Occurrence of Peak Abduction Moment	43.80 ± 26.97	44.10 ± 28.42	51.07 ± 31.36	40.12 ± 24.74

The same 13 dependant variables were tested against the two main effects and one interaction. The descent trials had more variables that were significantly different than did the ascent trials.

Significant differences were found for gender in VL average EMG ($F = 12.94$, $p = 0.001$), VM average EMG ($F = 14.03$, $p = 0.0006$), VL occurrence of peak EMG ($F = 13.86$, $p = 0.0007$), BF occurrence of peak EMG ($F = 13.01$, $p = 0.0009$), VL peak EMG ($F = 17.68$, $p = 0.0004$), VM peak EMG ($F = 7.08$, $p = 0.0116$), BF peak EMG ($F = 4.25$, $p = 0.0466$) and peak knee abduction moment ($F = 6.01$, $p = 0.0192$). Peak knee extension moment was close to having significant differences ($F = 4.04$, $p = 0.0521$).

The BF occurrence of peak EMG was significant for gender as mentioned above. It was also significant for q-angle ($F = 6.46$, $p = 0.0155$).

For the variables significant for gender, all mean values were higher for females, which the exception of peak knee abduction moment. This value was lower for females. All graphs showing the flexion/extension angles for the knee and the flexion/extension and adduction/abduction moments for the knee in the time-domain can be found in Appendices 2-4. The means of the significant gender variables for descent can be seen below: average and peak EMG (Figure 4.6), occurrence of peak EMG (Figure 4.7), and abduction moments (Figure 4.8). Knee extension moments have been included in this graph as well.

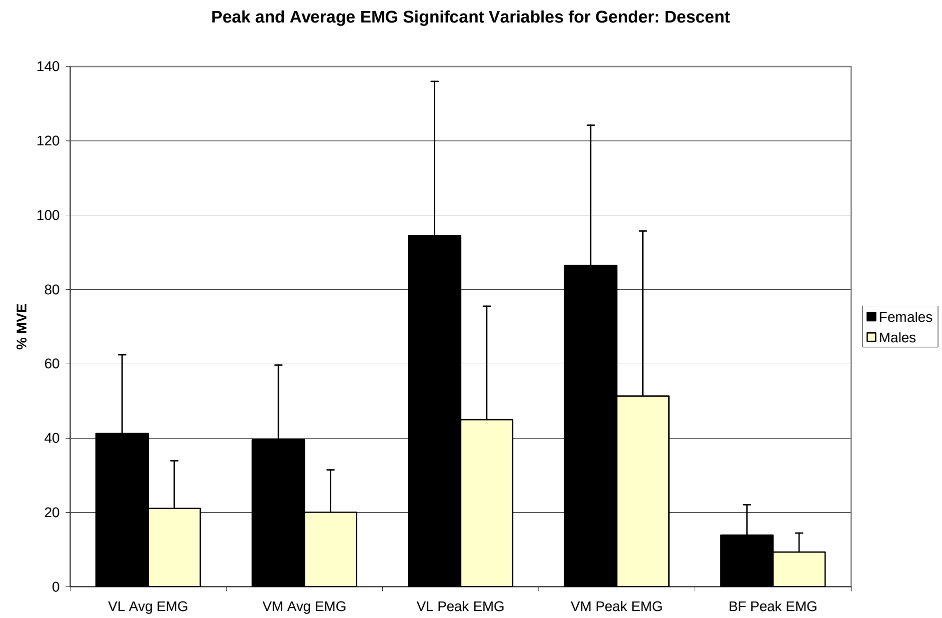


Figure 4.6*: The mean values for peak EMG and average EMG for the Vastus Lateralis (VL), Vastus Medialis (VM) and Biceps Femoris (BF) between genders. The error bars represent the '+' Standard Deviation (SD) within each variable. They are represented in terms of '% Maximum Voluntary Exertion' (MVE).*

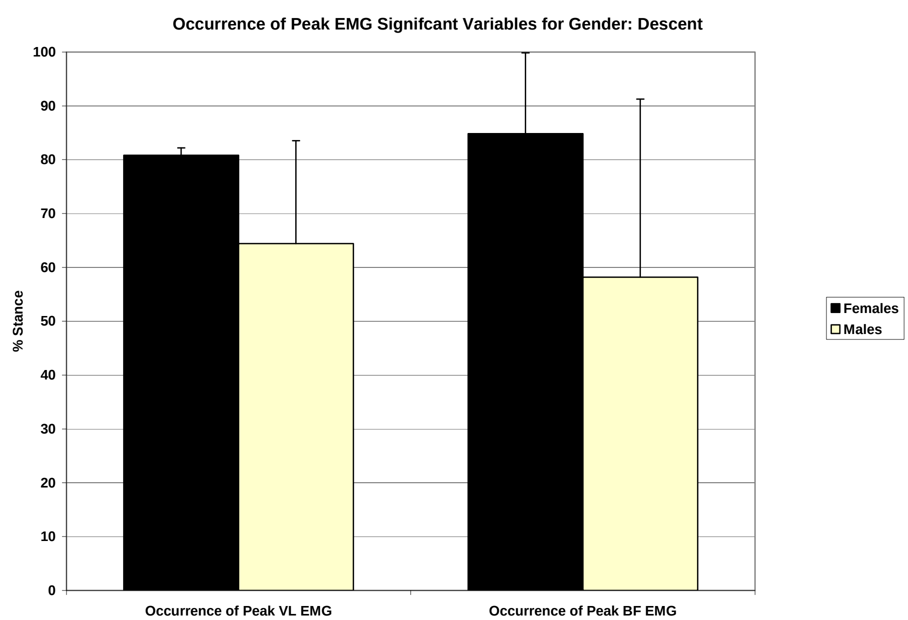


Figure 4.7*: The mean values for occurrence of peak for the Vastus Lateralis (VL) and Biceps Femoris (BF) between genders. The error bars represent the '+' Standard Deviation (SD) within each variable. They are represented in terms of '% Stance.*

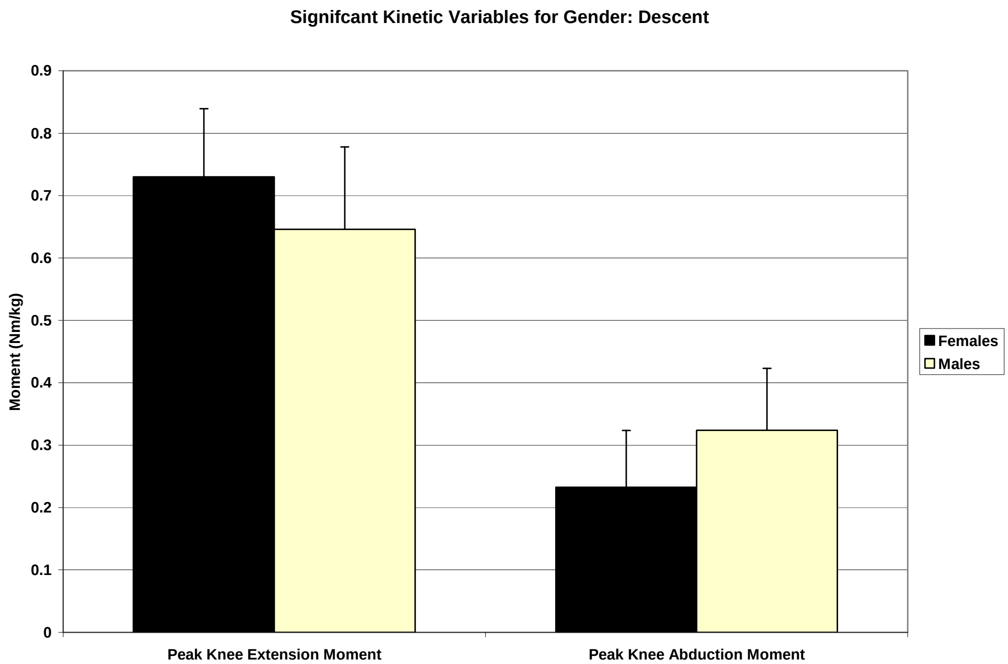


Figure 4.8*: The mean values for peak knee extension and abduction moments for stair descent between genders. The error bars represent the '+' Standard Deviation (SD) within each variable. The units are in Nm/kg.*

4.3 Discussion

4.3.1 Q-Angle

It was hypothesized that excessive q-angles would have increased quadriceps activity and increased knee moments. Only one of the 13 variables for descent and two of the 13 for ascent were significant for q-angle: biceps femoris average EMG and occurrence of peak knee abduction moment for stair ascent and occurrence of peak biceps femoris EMG for stair descent.

It can be seen from Table 4.3, that the q-angle values are similar to the values published by other researchers. Some studies had average values ranging from 9.5 - 18.4° (Livingston and Mandigo, 1999; Lathinghouse and Trimble, 2000; Kuhn et al., 2002; Shultz et al., 2006). Most of these studies used a self-selected normal stance or a supine position when measuring the q-angle. When looking at specific studies that used a Romberg stance, their values averaged 5.9° and 6.3° for the left and right leg of males and 9.7° and 10.1° for the left and right leg of females (Byl et al., 2000). The Romberg stance was chosen as this stance demonstrated the best representation of the quadriceps angle in a study looking at the change in q-angle with different stances (Livingston and Spaulding, 2002).

Another aspect of this study which differed from most of the others reviewed is the q-angles were collected using a motion capture system. Only two of the q-angle studies which have been reviewed in this paper used a motion capture system to collect the data. The remaining studies used a goniometer or x-ray. The studies which did use motion capture had similar average q-angle values (10.4 - 17.2 = left leg, 9.5 - 16.1 =

right leg) to this study (Livingston and Spaulding, 2002; Livingston and Mandigo, 1999; France and Nester, 2001).

The one study that was most comparable to the collection method used here, which was Romberg stance and motion capture, had similar ranges (1-25°) and average right and left leg q-angles (12.7° and 16.1°) to this study (Livingston and Spaulding, 2002). Our range was consistent with the aforementioned with the exception of subject 4 in the female group. This participant was overweight and not overly active. One study found that BMI was correlated with the malalignment in those with varus knees (Sharma et al., 2000). She did not report any known conditions or symptoms in the Knee Health Survey (Appendix 6). It can also be seen that for male subject 6, 8 and 9 and female subject 4, the right and left q-angle values differ by about 10°. The study by Livingston, et. al., (2002), saw differences between legs ranging from 8°-10.3°, which was one of the first studies to show limb asymmetry. Only recently has it become more common for studies to take bilateral measures. Many past studies assumed similar values for both legs (Horton and Hall, 1989; Woodland and Francis, 1992; Livingston and Spaulding, 1999; Aglietti et al., 1983; France and Nester, 2001; Lathinghouse and Trimble, 2000).

With respect to the q-angle effect on knee mechanics in stair ascent and descent, only three variables were significant. A main effect was found for biceps femoris average EMG and knee abduction occurrence of peak moment for stair ascent (Figure 4.9 & 4.10) and biceps femoris occurrence of peak EMG for stair descent (Figure 4.11). No significant interaction effects were found.

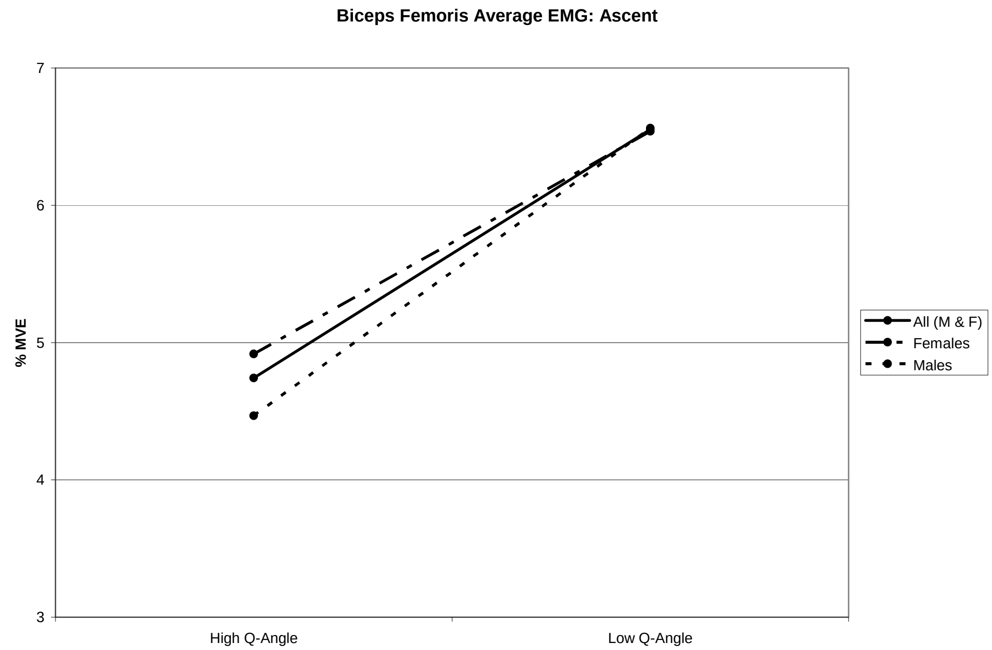


Figure 4.9: *The mean values average biceps femoris EMG for stair ascent. They are represented in terms of percent (%) of the maximum voluntary exertion (MVE). Females and Males both have n = 20. All (M&F) have n = 40 (high = 18, low =22).*

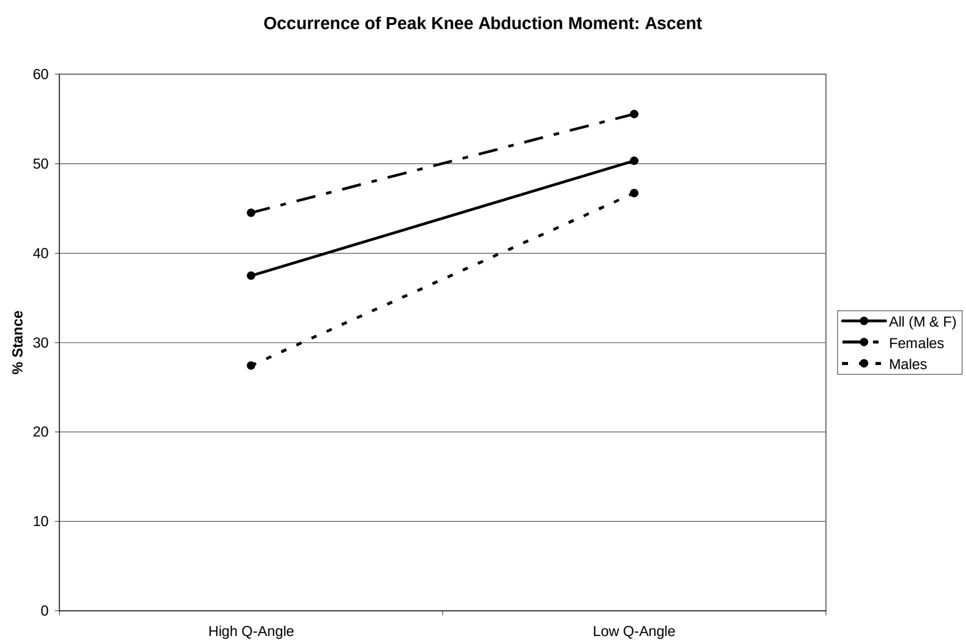


Figure 4.10: *The mean values for occurrence of peak knee abduction moment for stair ascent. They are represented in terms 0-100% Stance (heel strike to toe-off). Females and males both have n = 20. All (M&F) have n = 40 (high = 18, low =22).*

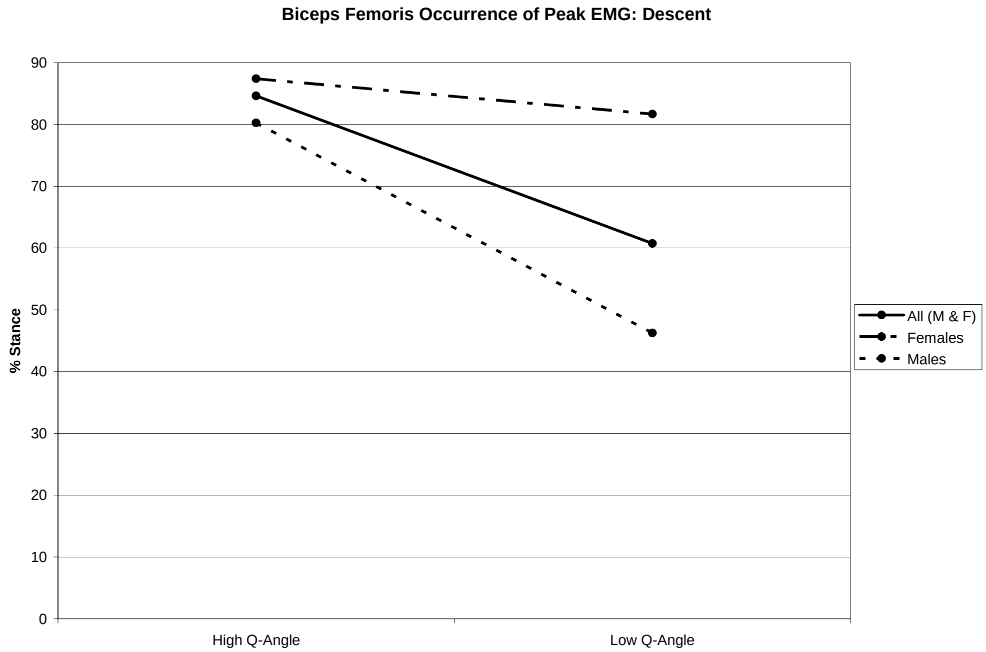


Figure 4.11*: The mean values for occurrence of peak biceps femoris EMG for stair descent. They are represented in terms 0-100% of foot contact with the force plate (heel strike to toe-off). Females and Males both have n = 20. All (M&F) have n = 40 (high = 18, low =22).*

As for the lack of significant effect of q-angle on knee mechanics and muscle activity, a few factors can be discussed. The q-angle for each leg was divided into 'high' (excessive) and 'low' (normal) groups. The excessive value was set at ≥15° for both males and females. This was a difficult decision as there is a lack of consensus in the literature in terms of an appropriate cutoff for an excessive q-angle. This value was chosen as this was the more popular trend in the studies examined (Horton and Hall, 1989; Livingston, 1998; Greene et al., 2001; Byl et al., 2000). In addition to this, one study was found which did separate q-angles into high and low groups and used 15° as the cutoff for both males and females (Heiderscheit et al., 1999). Recently, values between 8° - 10° for men and up to 15° for women are considered normal and that values above this can be problematic (Livingston and Spaulding, 2002; Greene et al., 2001; Livingston, 1998; Boden et al., 1997). Other excessive values were published for females as well, which considered >20° excessive (Csintalan et al., 2002; Horton and Hall, 1989). There was only one female with a q-angle above 20° in this study. It is suggested for future studies that multiple measures be made and an average taken before rating a participant as excessive or normal. Also, they should be required to walk around and reposition themselves to account for variability in stance.

However, there were a few trends seen for q-angle that did not reach statistical significance. In Figure 4.12, for 'all q-angles' and 'male' q-angles, there appears to be a difference between high and low in peak extension moment for stair descent, but females show no difference at all. A similar situation can be seen in Figure 4.13, where there appears to be a difference between high and low q-angle for males for knee abduction moment but when generalized to the entire data set, this difference is reduced. If a more ideal cutoff was in place with more balanced numbers in each group, the q-angle could become significant.

Although there was no statistically significant interaction effect found, the occurrence of peak biceps femoris EMG had an $F = 3.47$ and $p = 0.071$ for the interaction. When comparing the biceps femoris average EMG plot (Figure 4.9), the occurrence of biceps femoris plot (Figure 4.11) and the knee abduction plot (Figure 4.13) it can be seen that the high q-angle distinction seems to be affecting males more than females. It appears as though males might be affected by q-angle whereas females are not. This could be due to the natural anatomical variation in females or that the excessive cutoff was not appropriate. Knee abduction has been associated with an increased risk of developing medial compartment tibio-femoral osteoarthritis (Sharma et al., 1998; Messier et al., 1992). Therefore, high q-angles may predispose males to medial tibiofemoral osteoarthritis. This may affect females as well if a more appropriate excessive cutoff was established.

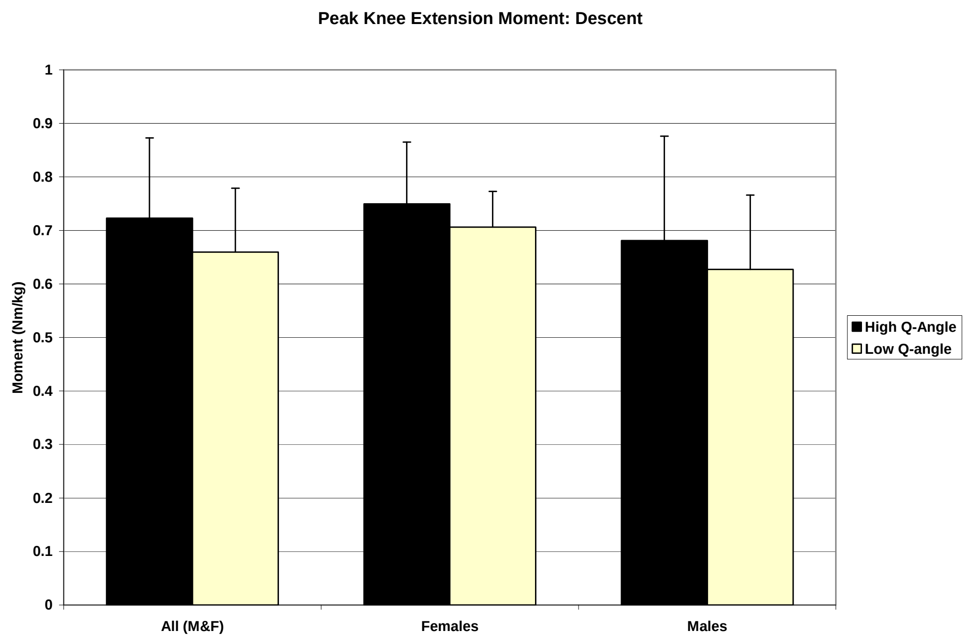


***Figure 4.12**: The mean values for peak knee extension moment for stair descent. The error bars represent the '+' Standard Deviation (SD) within each variable. They are represented in terms of maximum moment achieved (Nm/kg). Females and Males both have n = 20. All (M&F) have n = 40 (high = 18, low =22).*

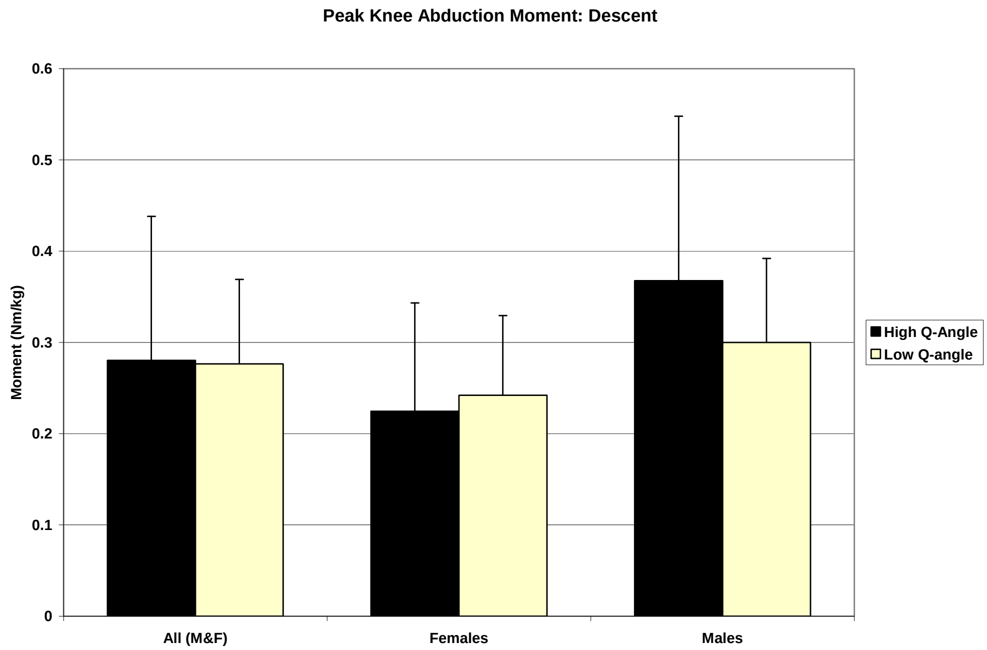


Figure 4.13*: The mean values for peak knee abduction moment for stair descent. The error bars represent the '+' Standard Deviation (SD) within each variable. They are represented in terms of maximum moment achieved (Nm/kg). Females and Males both have n = 20. All (M&F) have n = 40 (high = 18, low =22).*

During stair ascent there was a delayed occurrence of the peak abduction moment seen for the high q-angle group. When ascending the stairs, the peak angles are occurring later in the stance (~80%). With the peak abduction moment also happening later in the stance for the high q-angle group, the forces could be more damaging. It has been stated that when the knee is flexed more than 15° both the contact and shear loads rapidly increase (Taylor et al., 2004). The abduction moment would greatly increase the medio-lateral shear loads, when the tibio-femoral contact area is decreased.

There were also fewer people in the 'high' male group than was anticipated. There were only 7 of 20 subjects in the male group that were classified as having a high q-angle. Therefore, with the low numbers, significance that could have existed with greater numbers may have been concealed. The large standard deviation in Figure 4.12 seen in males could also be due to the small ‘n’ in the high group.

4.3.2 Gender

It was hypothesized that for gender, females would have higher peak moments as well as higher muscle activity. Females had increased vastus lateralis and medialis peak EMG and average EMG for stair ascent. For the muscle activity during descent, females demonstrated higher average EMG for the two quadriceps muscles (vastus lateralis and medialis), higher peak activations for all three muscles and delayed occurrence of peak EMG for vastus lateralis and biceps femoris. The gender differences for kinetics in stair descent exhibited that females had a decreased peak knee abduction moment. There was also a trend seen that females had a greater knee extension moment although not statistically significant.

The reduced number of significant findings for stair ascent is not unexpected. Andriacchi, *et al.,* (1980) found that the extension moments during stair ascent were decreased compared to those for descent. It was also found that the vertical ground reaction force was higher for the descent than the ascent which would increase the moment at the knee (McFadyen and Winter, 1988). This could be due to the fact that ascent is less demanding as it involves concentric contractions controlling the movement versus eccentric contractions which are more prevalent in descent (McFadyen and Winter, 1988).

During descent, the knee extension moment increases as the person lands on the step below. As the person is unweighting the contralateral limb to prepare for push off, the extension moment increases until stability is reached. The moments will increase as the body moves forward. The flexion moments at the start and finish represent the reversal of limb motion.

4.3.2.1 Kinetics

Due to there being no significant findings for kinetics in stair ascent, this section will discuss only stair descent. The one significant kinetic finding was decreased knee abduction moment for females. There was also a trend seen with an increased knee extension moment in females. One study looking at stair-climbing with OA patients found increased knee extension moments for females during walking, stair ascent and stair descent, with only the ascent being significant (Hughes et al., 2000). This was the only study reporting on gender differences in knee mechanics during stair-climbing.

As for the maximum mean knee extension moments reported here, they compare well to some in the literature. Protopapadaki *et. al.* (2007) exhibited knee extension moments at 0.51 Nm/kg, Andriacchi, *et al.* (1980) reported knee extension moments at 0.76 Nm/kg, Kowalk *et al.,* (1996) had values at 0.92 Nm/kg and Costigan, *et al.,* (2002) with moments slightly higher at 1.16 Nm/kg. The average peak value for females in this study was 0.73 Nm/kg and 0.65 Nm/kg for males.

The increased knee extension moment in females can be attributed to the increased knee flexion angles which can be seen in Appendix 3. The increased knee flexion angles are a function of height. Women are shorter and therefore have higher peak angles. In this study, females had an average knee angle of 79.9° and 99.7° during ascent and descent respectively. Males had angles at 76.6° for ascent and 96.3° for descent. Other stair climbing studies have also reported women as having higher peak flexion angles (Livingston et al., 1991; Hughes et al., 2000). An investigation involving knee height and knee pain, found that once knee height was normalized to total body height, the gender difference in knee angle disappeared (Hunter et al., 2005). As can be seen in the Figure 4.14 and 4.16, the knee angles differ for men and women between 20% and 60% of stance during ascent and 30% to 85% during descent. When looking at the knee moment profiles (Figure 4.15 and 4.17), peak knee moments during ascent occur roughly at 20% of stance and at 80% stance during descent. These peak moments are occurring within stance when the knee angles differ between genders. This would lend support to the differences in knee angle as one possible explanation for the increased knee extension moment and quadriceps EMG seen in females. The increased knee extension moment in females would increase the loading in the knee joint, which increases the risk of developing osteoarthritis (Luepongsak et al., 2002).

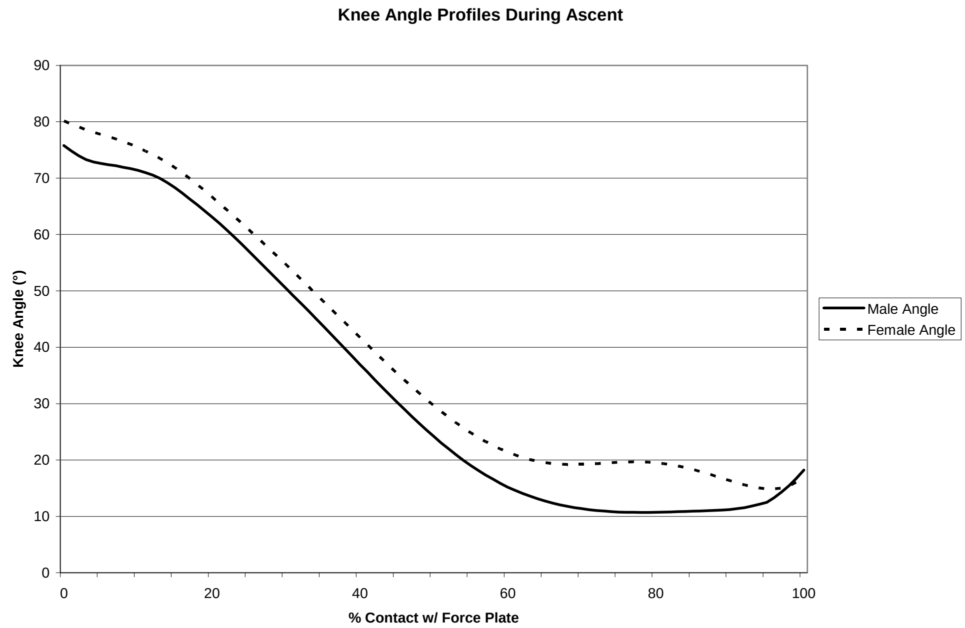


Figure 4.14*: Knee angle profiles during <u>ascent</u> for males and females. The knee angle profiles are represented in terms of % contact with the force plate. Knee Angles are in degrees (°).*

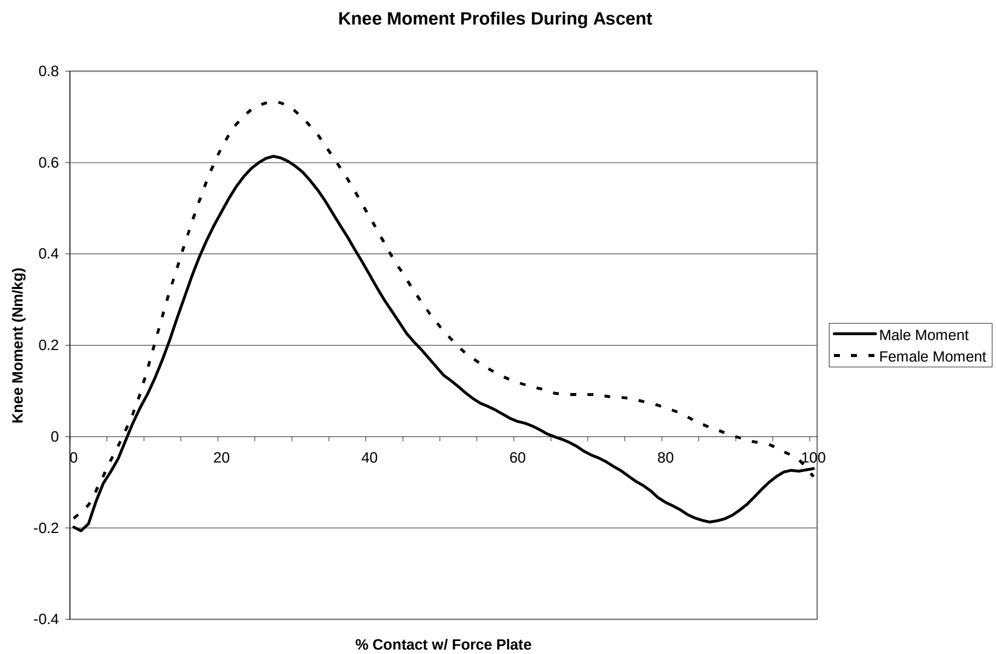


Figure 4.15*: Knee moment profiles during <u>ascent</u> for males and females. Knee moment profiles are represented in terms of % contact with the force plate. Knee moments are represented in Nm/kg.*

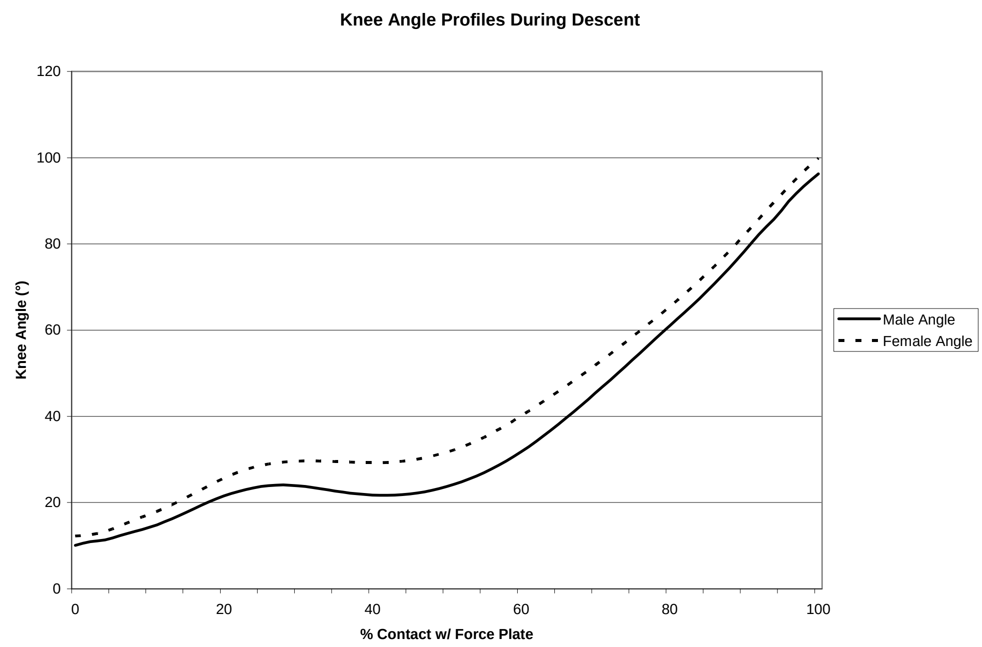


Figure 4.16: Knee angle profiles during <u>descent</u> for males and females. The knee angle profiles are represented in terms of % contact with the force plate. Knee Angles are in degrees (°).

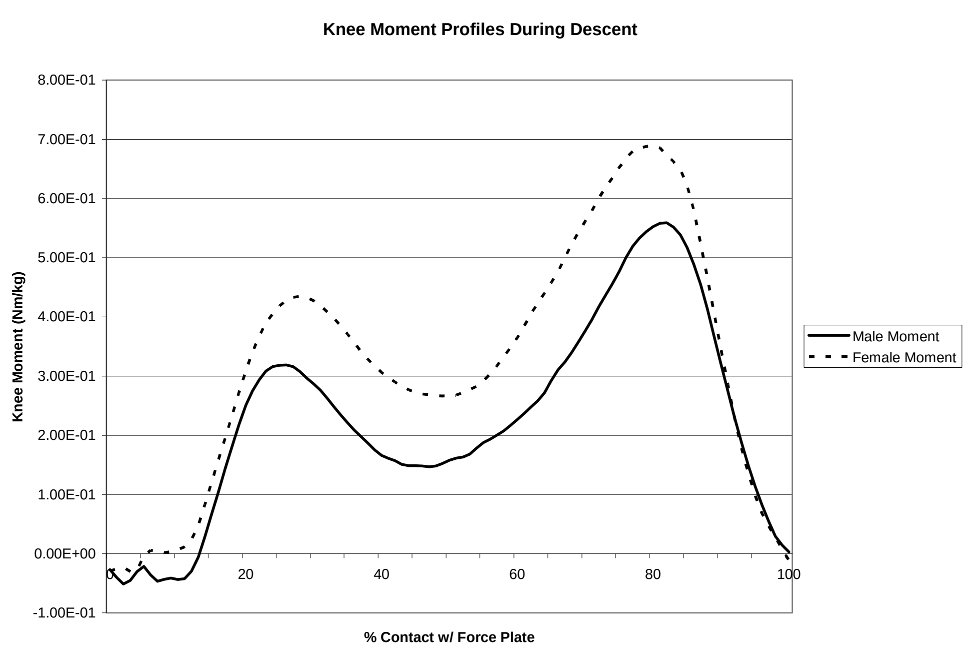


Figure 4.17: Knee moment profiles during <u>descent</u> for males and females. Knee moment profiles are represented in terms of % contact with the force plate. Knee moments are represented in Nm/kg.

When looking at cadence, on average females spent a slightly shorter amount of time in contact with the force plate (Table 4.5). It has been demonstrated that an increased cadence results in increased EMG activity in muscles of the thigh and leg (Yang and Winter, 1985; Zimmermann et al., 1994). Even though the average cadence in females was decreased, these means were very similar. Cadence varied in both groups with 6 females having increased cadence and the remaining four having slower cadence. Therefore, we cannot attribute the any of the kinetic or EMG findings to gender differences in cadence.

Table 4.5*: Time in Contact with the Force Plate. Highlighted boxes represent values that were higher for females, although not tested significantly.*

	Time in Contact with the Force Plate (s)	
	Males	**Females**
1	0.91	0.77
2	0.87	0.93
3	1.01	0.67
4	0.73	1.00
5	0.89	1.02
6	1.09	0.89
7	0.87	0.89
8	0.89	0.85
9	1.10	0.78
10	0.88	0.83
Mean	0.92	0.86

The critical finding in this study is the knee abduction moment. Currently there are no published gender differences for abduction moments during stair ascent or descent. Abduction moments are published less frequently as it is not the primary plane of motion and therefore has increased error. Due to the fact that our model used 2 sets of IReds to identify the knee joint rather that 1 set at the joint line, it was felt our data would be accurate. The abduction values in this study averaged 0.23 Nm/kg and 0.32 Nm/kg for females and males respectively. When comparing the abduction moments seen here, published values averaged 0.42, 0.46 and 0.62 Nm/kg (Costigan et al., 2002; Andriacchi et al., 1980; Kowalk et al., 1996). Our moments were slightly lower, but the three referenced studies had the force plate embedded into the ground under the steps, whereas ours was the second step. Also, their subjects wore shoes and one study used all males. The marker sets used also differed in that one study used only one marker at the midline of the knee (Andriacchi et al., 1980) and the other two studies used the fibular head as opposed to the tibial plateau used here (Costigan et al., 2002; Kowalk et al., 1996).

The decreased knee abduction moment in females could be explained by the fact that women naturally have a greater adduction angle at the hip due to a greater hip width to femur length ratio (Horton and Hall, 1989; Ferber et al., 2003). This increased hip adduction angle could lead to a decreased step width. A decreased step width when descending stairs would require less muscular effort to maintain the body's center of mass, therefore requiring less muscle activity. It was found that during stair descent, the center of mass shifted up to 4.2cm laterally, which would result in the knee abducting (i.e. internal abduction moment) (Zachazewski et al., 1993). Again, there is only one study which has looked at gender effects on stair-climbing and it looked only at the sagittal plane in elderly with knee OA (Hughes et al., 2000); therefore we have nothing with which to compare.

4.3.2.2 Electromyography (EMG)

The majority of the findings in this study were involving EMG, and consistently involved the same muscles. For both stair ascent and descent, females had significantly higher average EMG values and peak EMG values. This again comes back to the increased knee flexion angle and increased knee extension moment, seen in females above, which would both require higher quadriceps activation. Higher average quadriceps activity in females have been reported for other tasks such as a side-step cutting task (Sigward and Powers, 2006).

The increased quadriceps activation in stair descent could be explained by the decreased abduction moment. Females may be controlling knee mechanics by activating muscles differently than men. Women could be compensating for decreased abductor activation by increasing their quadriceps activation. The majority of the abduction that occurs at the knee is due to abduction occurring at the hip. Both the gluteus medius and tensor fascia latae (TFL) are hip abductors; with TFL also working as a hip flexor. The quadriceps function mainly as a knee extensor but also a hip flexor. Therefore if the abductors are weak, specifically the TFL, then the quadriceps muscles would have increased activation (Marieb, 2001).

There was a significant difference found for the biceps femoris between males and females but only for the descent. Sigward, *et. al.,* (2006) found increased biceps femoris activity in females for a side-step cutting task. The absence of hamstring significance during ascent could be due to the nature of the muscle activity. Ascending stairs involves knee extension which requires the quadriceps to contract concentrically to extend the leg (McFadyen and Winter, 1988). The hamstrings would be activating to support the knee and extend the hip therefore, having reduced activity as it is not the primary muscle group. During descent the hamstring muscles are working eccentrically to support the hip, which would increase the moment acting on the knee as it is a two joint muscle. Men and women contract much differently eccentrically than they do concentrically as demonstrated by Colliander *et. al.*,(1989). They demonstrated that women increased their hamstring eccentric peak torque as a function of increased angular velocity, where the men showed no change in their peak torque with the increased velocity. During descent, with the assistance of gravity, the angular velocity of the knee could be increased as subjects would descend more quickly than they would ascend.

These findings also explain why the moments were greater for males and females when EMG activity was decreased. One would think that if muscle activity is decreased, then the pull of the muscle acting on the joint would lead to a decreased moment. Eccentric muscle activity produces higher peak torques (Colliander and Tesch, 1989); therefore muscle activity could be reduced but the moment would still increase.

Finally, why was the occurrence of peak EMG for the biceps femoris and vastus lateralis delayed for females during stair descent (Figure 4.6)? And why was average biceps femoris EMG and occurrence of peak knee abduction moment for ascent and the biceps femoris occurrence of peak EMG for descent delayed for the high q-angle group (Figure 4.9-4.11)? Excessive q-angles have been linked to patellofemoral pain syndrome (Livingston and Mandigo, 1999; Tomsich et al., 1996). It has been reported that people suffering from patellofemoral pain exhibited trends of delayed vastus lateralis activation (Crossley et al., 2004). This would explain the delayed VL onset for the high q-angle group during stair ascent. Only the vasti were examined in this study; therefore we can not compare the biceps femoris. However, if there is delayed muscle activity in the quadriceps, this could carry over to the musculature of the entire thigh.

An alternative explanation involves the increased peak muscle activity and the delayed occurrence of peaks. When descending, subjects in these two groups could be using less muscle activity, ‘falling’ down to the next step at first then catching themselves creating higher peaks and taking longer to get to the peaks due to the muscles not being activated until later in the movement. This delayed firing in females could also be due to the larger range of motion (ROM) in females due to differences in height.

The gender differences found in the occurrence of peak activation levels can be partially explained by fiber type. It has been demonstrated that females have a higher percentage of type 1 fibers (slow twitch) compared to men (Mannion et al., 1998; Simoneau and Bouchard, 1989). These studies looked at the vastus lateralis and back extensor muscles. However, this trend could be generalized to the muscles of the thigh. If women have a higher percentage of slow twitch fibers in the biceps femoris, this would result in the delayed occurrence of peak exhibited here.

There were many subjects that had peak EMG values above 100% of the maximal exertion (Figure 4.4). The protocol used in this study, stair-climbing, is a highly dynamic activity which mostly targets the quadriceps muscles. Higher activations would be expected when compared to walking. Maximum exertions are difficult to complete as subjects themselves have to gauge whether they are nearing or at their maximums. They may not be putting in full effort for the contraction. Additionally, the method used to elicit the contraction may not have been ideal. As can be seen from Figure 4.1, the subject sat on the edge of a platform with their lower legs hanging over the edge. When the researcher manually resisted the MVE, the subject was not secured down, and therefore could have been recruiting other muscles to stabilize which would reduce the quadriceps contraction. With respect to the hamstring muscle, many individuals experienced cramping when completing the hamstring MVE, which in following trials, would cause them to reduce their level of contraction.

An alternate explanation for contraction >100%, is involving the neoprene wraps used with the tracking plates from the motion analysis. The MVE trials were completed at the start of the testing session. Following this, the motion markers and tracking plates were placed on the subjects. When the wraps were applied, this could have increased the contact surface that the EMG electrodes had with the skin. This would therefore affect the gain of the EMG during the stair-climbing. Only the vastus lateralis and medialis had excessively high values. As can be seen in Appendix 5, data was collected to test the effect of the neoprene wraps on the peak EMG values during stair ascent and descent. There are noticeable differences; however, no trend is obvious. The differences seen could be due to natural variance between trials.

4.3.3 Limitations

The model approach used in this study was to collect kinetic and kinematic data which, incorporated with segment inertial properties, output the forces and moments acting about the knee. However, this method only accounts for the net internal moments and does not directly measure the contribution from individual muscles, making it difficult to quantify antagonist muscle actions and isolate the muscle(s) responsible for changes in internal moments (Kellis, 1998).

Another limitation of this study is the skin motion associated with the motion analysis markers on the joints. Marker displacement associated with skin motion during knee bending was reported as having an RMS between 2mm and 17mm (Sati et al., 1996). However, this number can double if the marker is mistakenly displaced by 2cm when landmarking the condyles of the knee (Sati et al., 1996). It has also been noted that there is an increase in skin motion as you get closer to the joint line (Sati et al., 1996); therefore, it is recommended that the marker be placed at the edge of the condyle, away from the joint line, to reduce additional motion.

One major limitation of the study involves the equipment and protocol. Following the application of all of the equipment, the subject had 36 motion wires and 6 EMG wires hanging from a belt secured around their waist, which housed 4 marker packs. Additionally, they wore very tightly fitting neoprene wraps around each thigh and shank. These had to be snug enough that they would not slip during movement. Many subjects complained of a spring reaction occurring when they bent their knees to go up and down the stairs. This could change their mechanics and muscle activity, which alters how the subject would naturally and normally ascend and descend the steps. It was also noted that once the wraps were secured around the thigh, the resting level of quadriceps activity increased, as if the quads were activating to keep the wraps in place. The vastus lateralis on both legs were the only channels affected. A better method of securing the tracking plates would be advised for future studies.

6. Conclusions

This study confirms that gender differences in knee mechanics do exist with stair ascent and stair descent. The differences are much greater with descent than with ascent due to the eccentric nature of lower limb function. However, the main purpose of this study was to determine if excessive and normal q-angles caused different knee kinetics and muscle activity during stair-climbing.

The q-angle had significant differences for three variables: increased vastus lateralis peak EMG and delayed occurrence of peak knee abduction moment for stair ascent and delayed occurrence of peak biceps femoris EMG for stair descent (Figures 4.9-4.11). There were trends that were apparent when looking at excessive q-angles separately between genders. It can be seen in Figures 4.12 and 4.13 that males with high q-angles demonstrated differences in knee abduction moment and in knee extension moment, while the women did not. It does appear that male knee mechanics and function are more affected by q-angle than females. These differences could become significant for females with a larger subject pool or a different excessive measure. Of the 40 limbs being analyzed, only 18 were considered high and 22 were normal. There also needs to be a consensus as to what q-angle value is considered problematic for both males and females.

As exhibited in this study, females had both higher q-angle values and higher peak and average EMG for the vastus lateralis and medialis. Due to the involvement of the quadriceps group in the findings, it warrants further investigation into the relationships between q-angle and knee joint function. It would also be recommended that with the difference seen in knee abduction moments as well as the delayed occurrence of EMG seen in the biceps femoris, hip mechanics should be included in future studies. It has been stated that increased internal hip rotation coupled with greater knee abduction, could result in a greater dynamic q-angle (Ferber et al., 2003).

In conclusion, the increased moments at the knee indicate a greater risk of developing tibio-femoral osteoarthritis. The gender findings in this study alone are significant as no studies to date have been published regarding stair ascent and descent in a healthy young population.

Reference List

Aglietti, P., Insall, J. N., Cerulli, G., (1983). Patellar Pain and Incongruence .1. Measurements of Incongruence. Clinical Orthopaedics and Related Research 217-224.

Andriacchi, T. P., Andersson, G. B., Fermier, R. W., Stern, D., Galante, J. O., (1980). A study of lower-limb mechanics during stair-climbing. J.Bone Joint Surg.Am. 62**,** 749-757.

Andriacchi, T. P., Andersson, G. B., Ortengren, R., Mikosz, R. P., (1984). A study of factors influencing muscle activity about the knee joint. J.Orthop.Res. 1**,** 266-275.

Bayraktar, B., Yucesir, I., Ozturk, A., Cakmak, A. K., Taskara, N., Kale, A., Demiryurek, D., Bayramoglu, A., Camlica, H., (2004). Change of quadriceps angle values with age and activity. Saudi Medical Journal 25**,** 756-760.

Boden, B. P., Pearsall, A. W., Garrett, W. E., Jr., Feagin, J. A., Jr., (1997). Patellofemoral Instability: Evaluation and Management. J.Am.Acad.Orthop.Surg. 5**,** 47-57.

Byl, T., Cole, J. A., Livingston, L. A., (2000). What determines the magnitude of the Q angle? A preliminary study of selected skeletal and muscular measures. Journal of Sport Rehabilitation 9**,** 26-34.

Cerejo, R., Dunlop, D. D., Cahue, S., Channin, D., Song, J., Sharma, L., (2002). The influence of alignment on risk of knee osteoarthritis progression according to baseline stage of disease. Arthritis and Rheumatism 46**,** 2632-2636.

Collen, F. M., Baer, G. D., Ashburn, A. M., (2005). Stepping onto a single step: a kinematic study. Physiother.Res.Int. 10**,** 81-92.

Colliander, E. B., Tesch, P. A., (1989). Bilateral Eccentric and Concentric Torque of Quadriceps and Hamstring Muscles in Females and Males. European Journal of Applied Physiology and Occupational Physiology 59**,** 227-232.

Cooper, C., Mcalindon, T., Coggon, D., Egger, P., Dieppe, P., (1994). Occupational Activity and Osteoarthritis of the Knee. Annals of the Rheumatic Diseases 53**,** 90-93.

Costigan, P. A., Deluzio, K. J., Wyss, U. P., (2002). Knee and hip kinetics during normal stair climbing. Gait & Posture 16**,** 31-37.

Crossley, K. M., Cowan, S. M., Bennell, K. L., McConnell, J., (2004). Knee flexion during stair ambulation is altered in individuals with patellofemoral pain. J.Orthop.Res. 22**,** 267-274.

Csintalan, R. P., Schulz, M. M., Woo, J., McMahon, P. J., Lee, T. Q., (2002). Gender differences in patellofemoral joint biomechanics. Clinical Orthopaedics and Related Research 260-269.

Escamilla, R. F., (2001). Knee biomechanics of the dynamic squat exercise. Medicine and Science in Sports and Exercise 33**,** 127-141.

Ferber, R., Davis, I. M., Williams, D. S., (2003). Gender differences in lower extremity mechanics during running. Clinical Biomechanics 18**,** 350-357.

Fleiss, J. L., (1986). Reliability of Measurement. Wiley & Sons, New York, NY.

France, L., Nester, C., (2001). Effect of errors in the identification of anatomical landmarks on the accuracy of Q angle values. Clinical Biomechanics 16**,** 710-713.

Gelber, A. C., Hochberg, M. C., Mead, L. A., Wang, N. Y., Wigley, F. M., Klag, M. J., (2000). Joint injury in young adults and risk for subsequent knee and hip osteoarthritis. Annals of Internal Medicine 133**,** 321-328.

Greene, C. C., Edwards, T. B., Wade, M. R., Carson, E. W., (2001). Reliability of the quadriceps angle measurement. Am.J.Knee Surg. 14**,** 97-103.

Grood, E. S., Stowers, S. F., Noyes, F. R., (1988). Limits of Movement in the Human Knee - Effect of Sectioning the Posterior Cruciate Ligament and Posterolateral Structures. Journal of Bone and Joint Surgery-American Volume 70A**,** 88-97.

Guccione, A. A., Felson, D. T., Anderson, J. J., Anthony, J. M., Zhang, Y. Q., Wilson, P. W. F., Kellyhayes, M., Wolf, P. A., Kreger, B. E., Kannel, W. B., (1994). The Effects of Specific Medical Conditions on the Functional Limitations of Elders in the Framingham-Study. American Journal of Public Health 84**,** 351-358.

Gyory, A. N., Chao, E. Y. S., Stauffer, R. N., (1976). Functional Evaluation of Normal and Pathologic Knees During Gait. Archives of Physical Medicine and Rehabilitation 57**,** 571-577.

Hamel, K. A., Cavanagh, P. R., (2004). Stair performance in people aged 75 and older. Journal of the American Geriatrics Society 52**,** 563-567.

Hamel, K. A., Okita, N., Bus, S. A., Cavanagh, P. R., (2005). A comparison of foot/ground interaction during stair negotiation and level walking in young and older women. Ergonomics 48**,** 1047-1056.

Hassan, B. S., Doherty, S. A., Mockett, S., Doherty, M., (2002). Effect of pain reduction on postural sway, proprioception, and quadriceps strength in subjects with knee osteoarthritis. Annals of the Rheumatic Diseases 61**,** 422-428.

Heiderscheit, B. C., Hamill, J., van Emmerik, R. E., (1999). Q-angle influences on the variability of lower extremity coordination during running. Med.Sci.Sports Exerc. 31**,** 1313-1319.

Hinman, R. S., Cowan, S. M., Crossley, K. M., Bennell, K. L., (2005). Age-related changes in electromyographic quadriceps activity during stair descent. Journal of Orthopaedic Research 23**,** 322-326.

Hood, V., Nicol, A., (2005). Biomechanics of Stair Descent in Older Adults.

Hortobagyi, T., Westerkamp, L., Beam, S., Moody, J., Garry, J., Holbert, D., DeVita, P., (2005). Altered hamstring-quadriceps muscle balance in patients with knee osteoarthritis. Clinical Biomechanics 20**,** 97-104.

Horton, M. G., Hall, T. L., (1989). Quadriceps Femoris Muscle Angle - Normal Values and Relationships with Gender and Selected Skeletal Measures. Physical Therapy 69**,** 897-901.

Hughes, C. A., Kaufman, K. R., Morrey, B. F., Morrey, M. A., An, K. N., (2000). Gender Differences in Knee Kinematics and Kinetics During Stair Climbing and Level Walking in Adults with Osteoarthritis of the Knee.

Hunter, D. J., Niu, J. B., Zhang, Y. Q., Nevitt, M. C., Xu, L., Lui, L. Y., Yu, W., Aliabadi, P., Buchanan, T. S., Felson, D. T., (2005). Knee height, knee pain, and knee osteoarthritis - The Beijing Osteoarthritis Study. Arthritis and Rheumatism 52**,** 1418-1423.

Huston, L. J., Greenfield, M. L. V. H., Wojtys, E. M., (2000). Anterior cruciate ligament injuries in the female athlete - Potential risk factors. Clinical Orthopaedics and Related Research 50-63.

Kaufman, K. R., Hughes, C., Morrey, B. F., Morrey, M., An, K. N., (2001). Gait characteristics of patients with knee osteoarthritis. Journal of Biomechanics 34**,** 907-915.

Kellis, E., (1998). Quantification of quadriceps and hamstring antagonist activity. Sports Medicine 25**,** 37-62.

Kowalk, D. L., Duncan, J. A., Vaughan, C. L., (1996). Abduction-adduction moments at the knee during stair ascent and descent. J.Biomech. 29**,** 383-388.

Kuhn, D. R., Yochum, T. R., Cherry, A. R., Rodgers, S. S., (2002). Immediate changes in the quadriceps femoris angle after insertion of an orthotic device. Journal of Manipulative and Physiological Therapeutics 25**,** 465-470.

Lathinghouse, L. H., Trimble, M. H., (2000). Effects of isometric quadriceps activation on the Q-angle in women before and after quadriceps exercise. Journal of Orthopaedic & Sports Physical Therapy 30**,** 211-216.

Livingston, L. A., (1998). The quadriceps angle: A review of the literature. Journal of Orthopaedic & Sports Physical Therapy 28**,** 105-109.

Livingston, L. A., Mandigo, J. L., (1999). Bilateral Q angle asymmetry and anterior knee pain syndrome. Clinical Biomechanics 14**,** 7-13.

Livingston, L. A., Spaulding, S. J., (1999). Three-Dimensional Quantification of Concomitant Changes in Knee Flexion-Extension and Quadriceps Angles.

Livingston, L. A., Spaulding, S. J., (2002). OPTOTRAK measurement of the quadriceps angle using standardized foot positions. Journal of Athletic Training 37**,** 252-255.

Livingston, L. A., Stevenson, J. M., Olney, S. J., (1991). Stairclimbing Kinematics on Stairs of Differing Dimensions. Archives of Physical Medicine and Rehabilitation 72**,** 398-402.

Luepongsak, N., Amin, S., Krebs, D. E., McGibbon, C. A., Felson, D., (2002). The contribution of type of daily activity to loading across the hip and knee joints in the elderly. Osteoarthritis.Cartilage. 10**,** 353-359.

Lyons, K., Perry, J., Gronley, J. K., Barnes, L., Antonelli, D., (1983). Timing and Relative Intensity of Hip Extensor and Abductor Muscle Action During Level and Stair Ambulation - An Emg Study. Physical Therapy 63**,** 1597-1605.

Mannion, A. F., Dumas, G. A., Stevenson, J. M., Cooper, R. G., (1998). The influence of muscle fiber size and type distribution on electromyographic measures of back muscle fatigability. Spine 23**,** 576-584.

Marieb, E. E., (2001). Human Anatomy and Physiology. Benjamin Cummings, New York, USA.

McFadyen, B. J., Winter, D. A., (1988). An Integrated Biomechanical Analysis of Normal Stair Ascent and Descent. Journal of Biomechanics 21**,** 733-744.

Messier, S. P., Loeser, R. F., Hoover, J. L., Semble, E. L., Wise, C. M., (1992). Osteoarthritis of the Knee - Effects on Gait, Strength, and Flexibility. Archives of Physical Medicine and Rehabilitation 73**,** 29-36.

Mizuno, Y., Kumagai, M., Mattessich, S. M., Elias, J. J., Ramrattan, N., Cosgarea, A. J., Chao, E. Y. S., (2001). Q-angle influences tibiofemoral and patellofemoral kinematics. Journal of Orthopaedic Research 19**,** 834-840.

Morrison, J. B., (1970). Mechanics of Knee Joint in Relation to Normal Walking. Journal of Biomechanics 3**,** 51-&.

Muller, R., Bisig, A., Kramers, A., Stussi, E., (1998). Influence of Stair Inclination on Muscle Activity in Normals. p. 32.

Okita, N., Cavanagh, P. R., (2001). Effects of aging on knee joint moment profiles during stair negotiation. p. 281.

Owings, T. M., Grabiner, M. D., (2004). Variability of step kinematics in young and older adults. Gait & Posture 20**,** 26-29.

Perotto, A. O., Delagi, E. F., Iazzetti, J., Morrison, D., (2005). Anatomical Guide for the Electromyographer: The Limbs and Trunk. Charles C. Thomas: Publisher Ltd., Springfield, Illinois.

Protopapadaki, A., Drechsler, W. I., Cramp, M. C., Coutts, F. J., Scott, O. M., (2007). Hip, knee, ankle kinematics and kinetics during stair ascent and descent in healthy young individuals. Clin.Biomech.(Bristol., Avon.) 22**,** 203-210.

Sati, M., Guise, J. A., Larouche, S., Drouin, G., (1996). Quantitative Assessment of Skin-Bone Movement of the Knee. The Knee 3**,** 121-138.

Sharma, L., Hurwitz, D. E., Thonar, E. J. M. A., Sum, J. A., Lenz, M. E., Dunlop, D. D., Schnitzer, T. J., Kirwan-Mellis, G., Andriacchi, T. P., (1998). Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis and Rheumatism 41**,** 1233-1240.

Sharma, L., Lou, C., Cahue, S., Dunlop, D. D., (2000). The mechanism of the effect of obesity in knee osteoarthritis: the mediating role of malalignment. Arthritis Rheum. 43**,** 568-575.

Shrout, P. E., Fleiss, J. L., (1979). Intraclass Correlations - Uses in Assessing Rater Reliability. Psychological Bulletin 86**,** 420-428.

Shultz, S. J., Nguyen, A. D., Windley, T. C., Kulas, A. S., Botic, T. L., Beynnon, B. D., (2006). Intratester and intertester reliability of clinical measures of lower extremity anatomic characteristics: Implications for multicenter studies. Clinical Journal of Sport Medicine 16**,** 155-161.

Sigward, S. M., Powers, C. M., (2006). The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting. Clin.Biomech.(Bristol., Avon.) 21**,** 41-48.

Siler, W. L., Jorgensen, A. L., Norris, R. A., (1997). Grasping the handrails during treadmill walking does not alter sagittal plane kinematics of walking. Archives of Physical Medicine and Rehabilitation 78**,** 393-398.

Simoneau, J. A., Bouchard, C., (1989). Human variation in skeletal muscle fiber-type proportion and enzyme activities. Am.J.Physiol 257**,** E567-E572.

Stauffer, R. N., Chao, E. Y. S., Gyory, A. N., (1977). Biomechanical Gait Analysis of Diseased Knee-Joint. Clinical Orthopaedics and Related Research 246-255.

Taylor, W. R., Heller, M. O., Bergmann, G., Duda, G. N., (2004). Tibio-femoral loading during human gait and stair climbing. Journal of Orthopaedic Research 22**,** 625-632.

Tomsich, D. A., Nitz, A. J., Threlkeld, A. J., Shapiro, R., (1996). Patellofemoral alignment: reliability. J.Orthop.Sports Phys.Ther. 23**,** 200-208.

Winter, D. A., (2005). Biomechanics and Motor Control of Human Movement. John Wiley & Sons INC..

Wojtys, E. M., Huston, L. J., Schock, H. J., Boylan, J. P., shton-Miller, J. A., (2003). Gender differences in muscular protection of the knee in torsion in size-matched athletes. Journal of Bone and Joint Surgery-American Volume 85A**,** 782-789.

Woodland, L. H., Francis, R. S., (1992). Parameters and Comparisons of the Quadriceps Angle of College-Aged Men and Women in the Supine and Standing Positions. American Journal of Sports Medicine 20**,** 208-211.

Yang, J. F., Winter, D. A., (1985). Surface Emg Profiles During Different Walking Cadences in Humans. Electroencephalography and Clinical Neurophysiology 60**,** 485-491.

Yu, B., Kienbacher, T., Growney, E. S., Johnson, M. E., An, K. N., (1997a). Reproducibility of the kinematics and kinetics of the lower extremity during normal stair-climbing. Journal of Orthopaedic Research 15**,** 348-352.

Yu, B., Stuart, M. J., Kienbacher, T., Growney, E. S., An, K. N., (1997b). Valgus-varus motion of the knee in normal level walking and stair climbing. Clinical Biomechanics 12**,** 286-293.

Zachazewski, J. E., Riley, P. O., Krebs, D. E., (1993). Biomechanical analysis of body mass transfer during stair ascent and descent of healthy subjects. J.Rehabil.Res.Dev. 30**,** 412-422.

Zhang, Y. Q., Hunter, D. J., Nevitt, M. C., Xu, L., Niu, J. B., Lui, L. Y., Yu, W., Aliabadi, P., Felson, D. T., (2004). Association of squatting with increased prevalence of radiographic tibiofemoral knee osteoarthritis - The Beijing Osteoarthritis study. Arthritis and Rheumatism 50**,** 1187-1192.

Zimmermann, C. L., Cook, T. M., Bravard, M. S., Hansen, M. M., Honomichl, R. T., Karns, S. T., Lammers, M. A., Steele, S. A., Yunker, L. K., Zebrowski, R. M., (1994). Effects of Stair-Stepping Exercise Direction and Cadence on Emg Activity of Selected Lower-Extremity Muscle Groups. Journal of Orthopaedic & Sports Physical Therapy 19**,** 173-180.

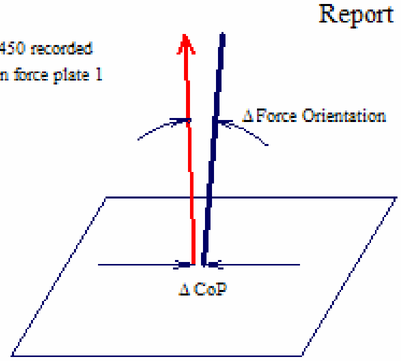
Appendix 1: The CalTester™ Test Report determining the amount of error introduced between the 3D Optotrak Certus™ motion capture system (Northern Digital Inc., Waterloo, ON) and the 6-channel force plate (Model: OR6-7-2000, AMTI, Watertown, MA)

Laboratory Calibration Test Report
University of Waterloo, Dept. of Kinesiology

Filename: c#001.c3d

Report created: 1 - August - 2007

Number of Frames: 351 of 450 recorded
Minimum Force (N): 40 on force plate 1
Number of Targets Used: 4



Report Format

Min (at xxx N) Mean ± 1 Std. Dev. Max (at xxx N)

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force at the time of the minimum and maximum value is indicated alongside the value.

Summary of Differences

Δ Force Orientation (°)= 1.2 ± 0.7 — 0.3 (at 61 N) — 3.3 (at 47 N)

Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the long axis of the device, as determined from the target data.

Δ CoP x (mm) = -1.9 ± 1.4 — -5.4 (at 54 N) — 1.6 (at 56 N)

Δ CoP y (mm) = 0.6 ± 0.1 — 0.5 (at 53 N) — 0.9 (at 48 N)

Δ CoP z (mm) = 5.8 ± 1.4 — 2.2 (at 54 N) — 10.7 (at 53 N)

Δ CoP x,y,z are the components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 0.0 mm above the force platform.)

Report generated by CalTester (© C-Motion Inc.)

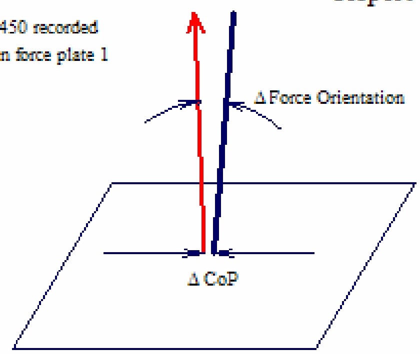
1.1: This demonstrates how much the mean centre of pressure changes (± standard deviation) within the range tested for corner #1 of the force plate. The amount of force exerted on the rod through the range of motion is also shown.

Laboratory Calibration Test Report
University of Waterloo, Dept. of Kinesiology

Filename: c#002.c3d

Report created: 1 - August - 2007

Number of Frames: 392 of 450 recorded
Minimum Force (N): 40 on force plate 1
Number of Targets Used: 4



Report Format

Min (at xxx N) Mean ± 1 Std. Dev. Max (at xxx N)

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force at the time of the minimum and maximum value is indicated alongside the value.

Summary of Differences

Δ Force Orientation (°)= 0.9 ± 0.4 — 0.2 (at 63 N) — 1.9 (at 59 N)

Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the long axis of the device, as determined from the target data.

Δ CoP x (mm) = 5.8 ± 1.5 — 1.7 (at 74 N) — 11.1 (at 50 N)

Δ CoP y (mm) = 0.6 ± 0.1 — 0.3 (at 56 N) — 0.8 (at 57 N)

Δ CoP z (mm) = 11.7 ± 1.6 — 8.0 (at 75 N) — 17.2 (at 50 N)

Δ CoP x,y,z are the components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 0.0 mm above the force platform.)

Report generated by CalTester (© C-Motion Inc.)

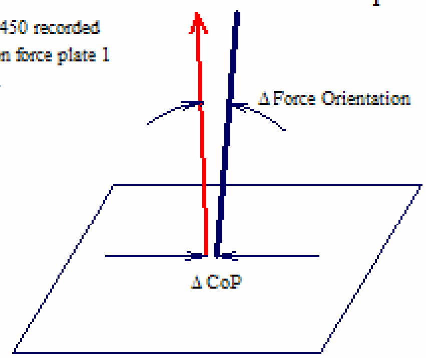
1.2: This demonstrates how much the mean centre of pressure changes (± standard deviation) within the range tested for corner #2 of the force plate. The amount of force exerted on the rod through the range of motion is also shown.

Laboratory Calibration Test Report
University of Waterloo, Dept. of Kinesiology

Filename: c#003.c3d

Report created: 1 - August - 2007

Number of Frames: 428 of 450 recorded
Minimum Force (N): 40 on force plate 1
Number of Targets Used: 4



Report Format

Min (at xxx N) Mean ± 1 Std. Dev. Max (at xxx N)

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force at the time of the minimum and maximum value is indicated alongside the value.

Summary of Differences

Δ Force Orientation (°)= 1.1 ± 0.5 — 0.1 (at 58 N) — 3.0 (at 48 N)

Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the long axis of the device, as determined from the target data.

Δ CoP x (mm) = 5.9 ± 1.6 — 2.4 (at 84 N) — 10.9 (at 40 N)

Δ CoP y (mm) = 0.6 ± 0.1 — 0.4 (at 56 N) — 1.1 (at 61 N)

Δ CoP z (mm) = 11.5 ± 1.8 — 7.4 (at 60 N) — 17.2 (at 49 N)

Δ CoP x,y,z are the components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 0.0 mm above the force platform.)

Report generated by CalTester (© C-Motion Inc.)

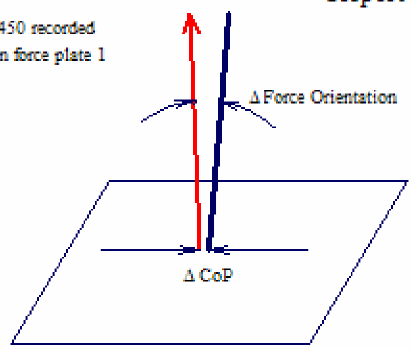
1.3: This demonstrates how much the mean centre of pressure changes (± standard deviation) within the range tested for corner #3 of the force plate. The amount of force exerted on the rod through the range of motion is also shown.

Laboratory Calibration Test Report
University of Waterloo, Dept. of Kinesiology

Filename: c#004.c3d

Report created: 1 - August - 2007

Number of Frames: 373 of 450 recorded
Minimum Force (N): 40 on force plate 1
Number of Targets Used: 4



Report Format

Min (at xxx N) Mean ± 1 Std. Dev. Max (at xxx N)

The mean, standard deviation, and range (minimum and maximum) of the report variables are provided on the graph and with numerical values. The applied force at the time of the minimum and maximum value is indicated alongside the value.

Summary of Differences

Δ Force Orientation (°)= 0.5 ± 0.3 — 0.0 (at 72 N) — 1.2 (at 62 N)

Δ Force Orientation is the angle between the orientation of the axial component of the applied force and the orientation of the long axis of the device, as determined from the target data.

Δ CoP x (mm) = -0.6 ± 1.0 — -3.9 (at 57 N) — 1.9 (at 61 N)

Δ CoP y (mm) = 0.6 ± 0.0 — 0.5 (at 61 N) — 0.7 (at 65 N)

Δ CoP z (mm) = 5.8 ± 1.0 — 3.5 (at 68 N) — 10.3 (at 44 N)

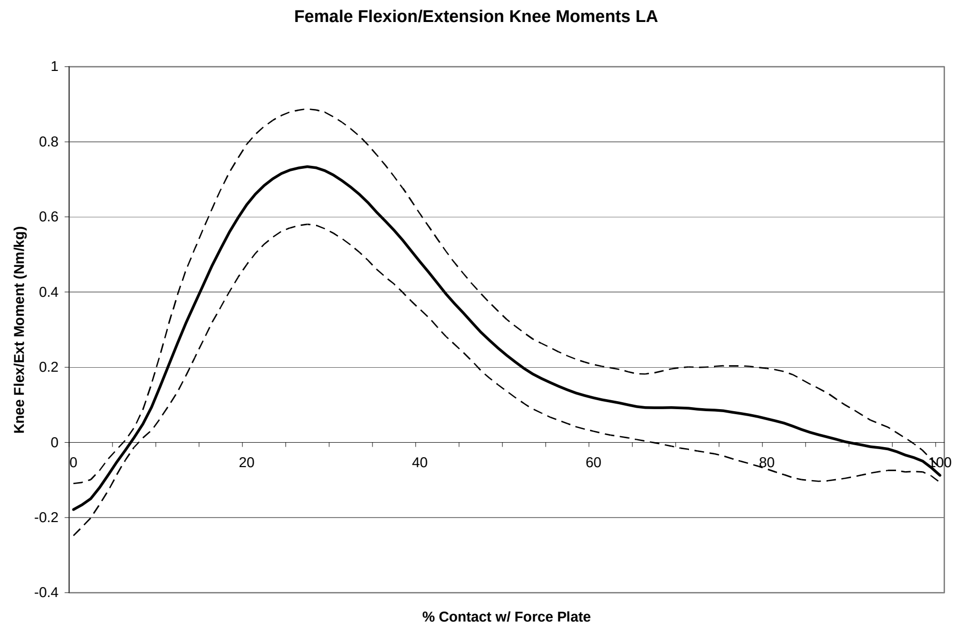
Δ CoP x,y,z are the components of the displacement vector between the Center of Pressure location measured by the force platform and the endpoint of the calibration testing device (adjusted for a vertical height of 0.0 mm above the force platform.)

Report generated by CalTester (© C-Motion Inc.)

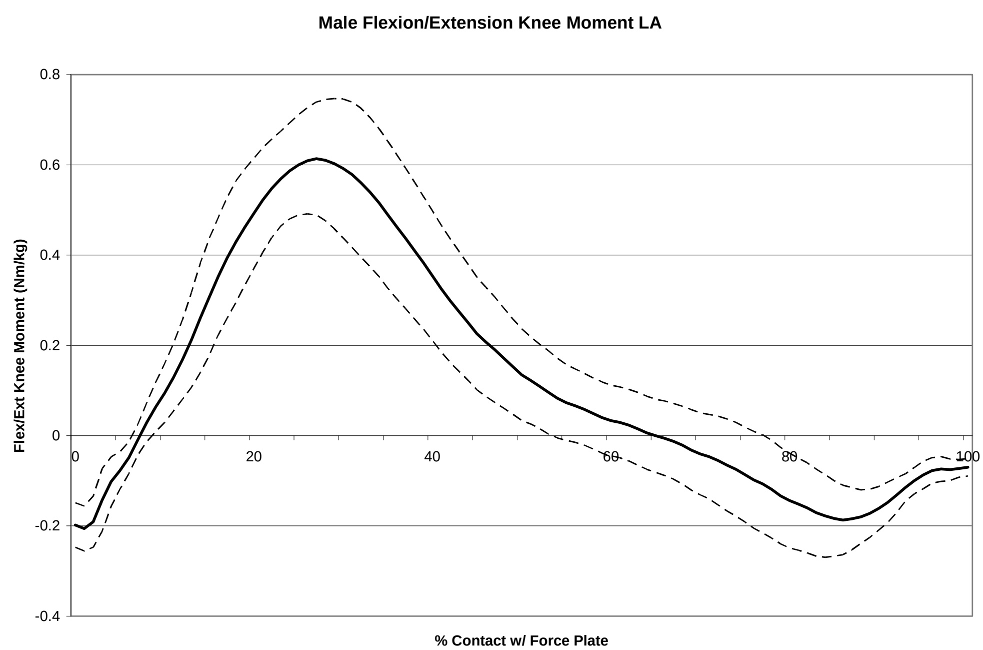
1.4: This demonstrates how much the mean centre of pressure changes (± standard deviation) within the range tested for corner #4 of the force plate. The amount of force exerted on the rod through the range of motion is also shown.

The largest difference was seen in 1.2. This 11mm difference would only alter the moment by 0.057 Nm/kg. This was calculated using data from Kingma et. al. (1996) who induced a 10mm translation.

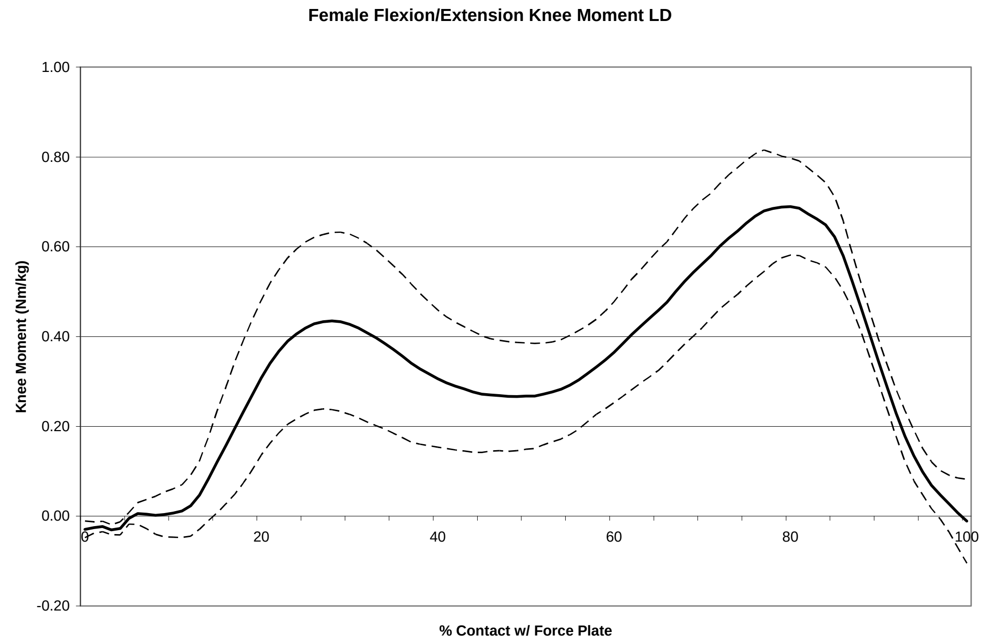
Appendix 2: Knee Extension Moments of Males and Females during Stair Ascent and Descent



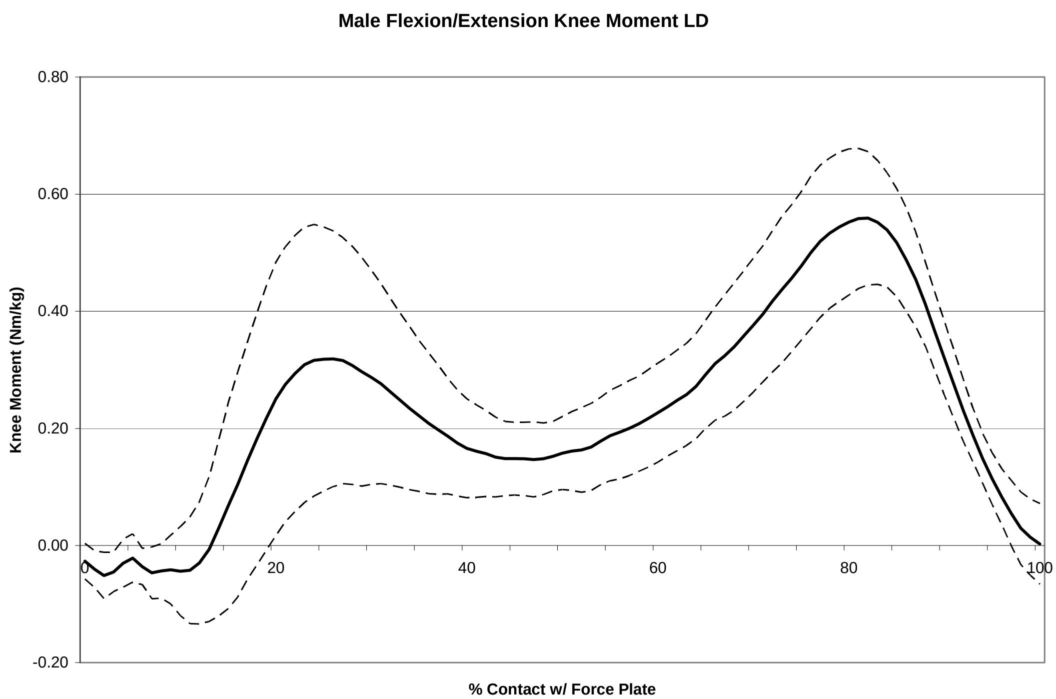
2.1: Knee moment profiles for the left leg of females during stair ascent. Positive ‘y’ signifies an extension moment and negative ‘y’ is flexion. The solid line represents the group mean while the dotted lines represent the $\pm$ Standard Deviation (SD). LA is left ascent.



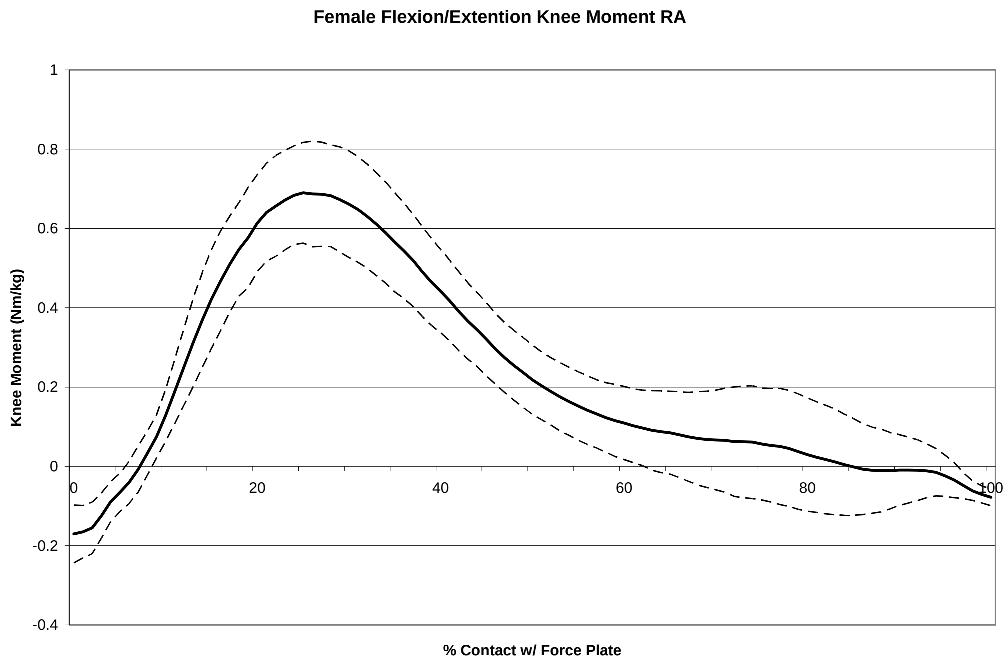
2.2: Knee moment profiles for the left leg of males during stair ascent. Positive ‘y’ signifies an extension moment and negative ‘y’ is flexion. The solid line represents the group mean while the dotted lines represent the $\pm$ Standard Deviation (SD). LA is left ascent.



2.3: Knee moment profiles for the left leg of females during stair descent. Positive ‘y’ signifies an extension moment and negative ‘y’ is flexion. The solid line represents the group mean while the dotted lines represent the <u>+</u> Standard Deviation (SD). LD is left descent.

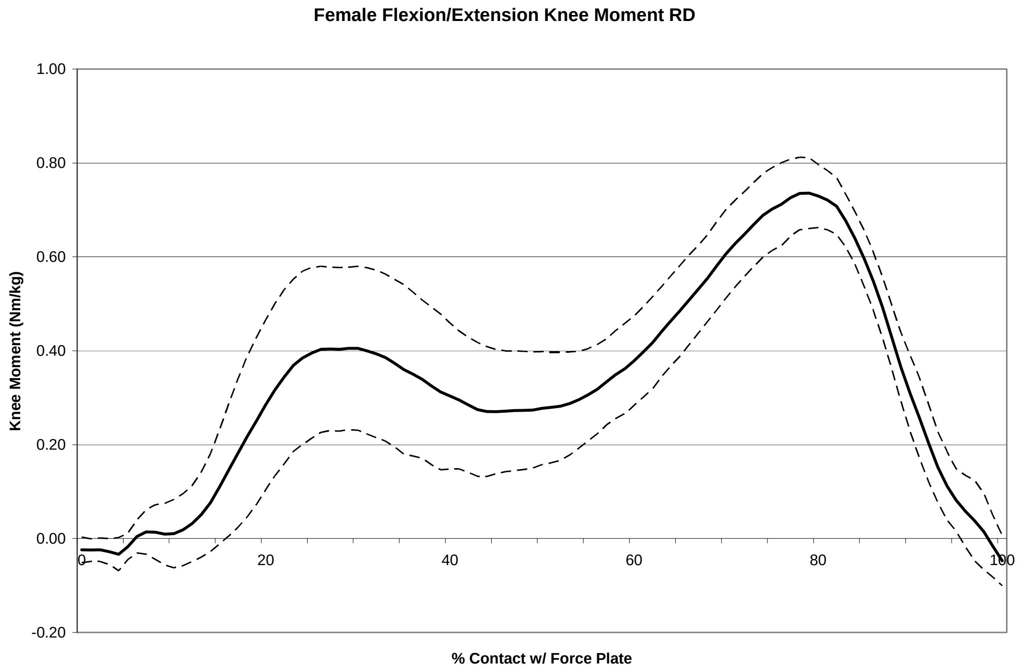


2.4: Knee moment profiles for the left leg of males during stair descent. Positive ‘y’ signifies an extension moment and negative ‘y’ is flexion. The solid line represents the group mean while the dotted lines represent the <u>+</u> Standard Deviation (SD). LD is left descent.



2.5: Knee moment profiles for the left right of females during stair ascent. Positive 'y' signifies an extension moment and negative 'y' is flexion. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). RA is right ascent.

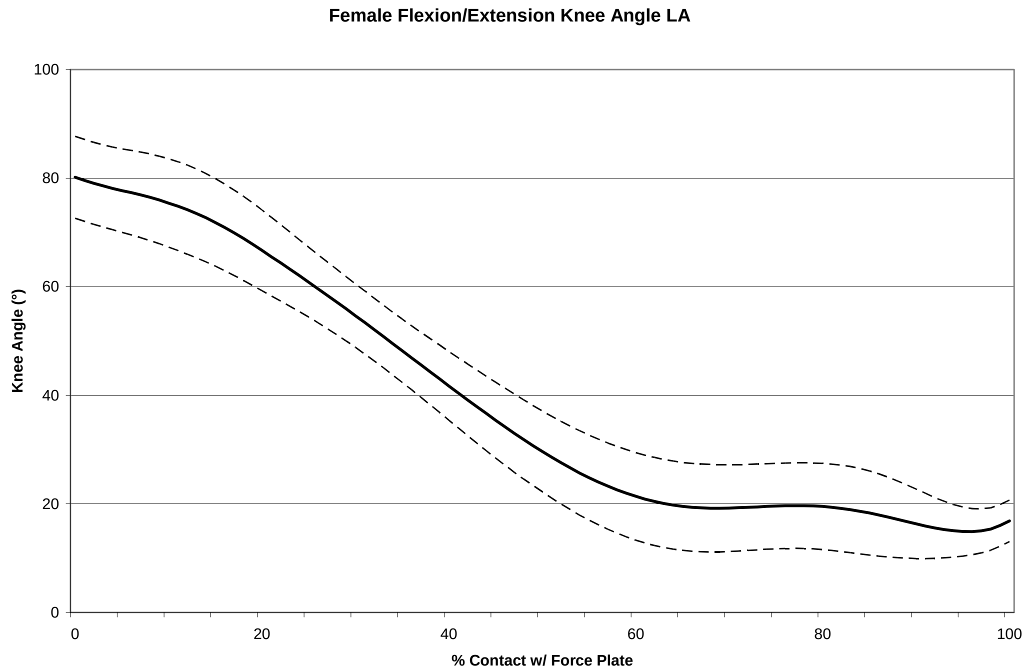
2.6: Knee moment profiles for the right leg of males during stair ascent. Positive 'y' signifies an extension moment and negative 'y' is flexion. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). RA is right ascent.



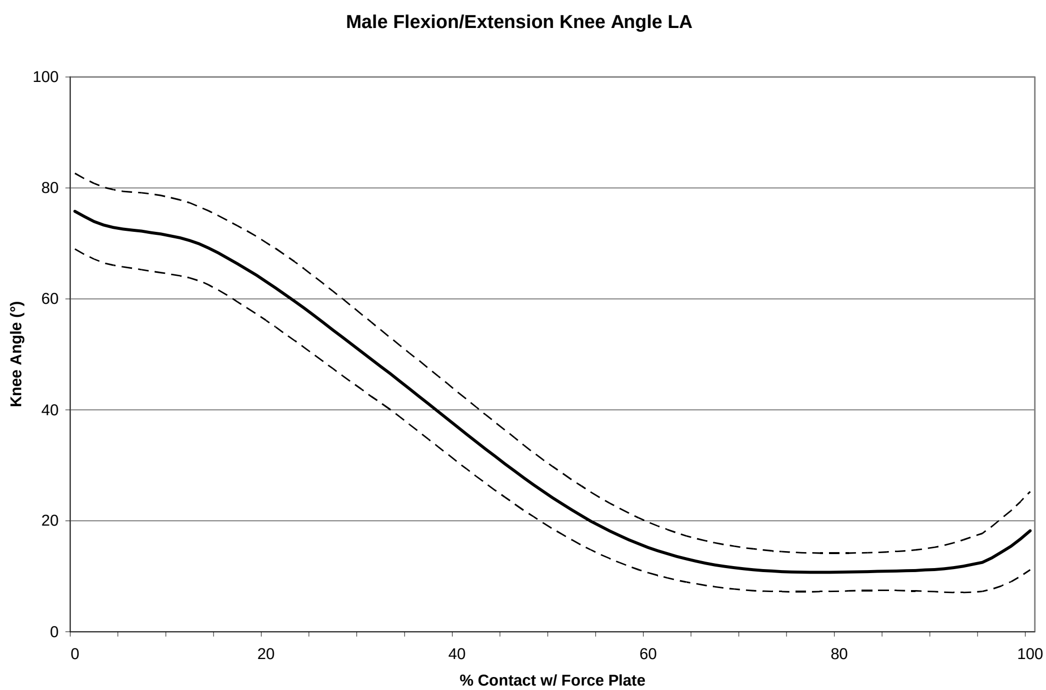
2.7: Knee moment profiles for the right leg of females during stair descent. Positive ‘y’ signifies an extension moment and negative ‘y’ is flexion. The solid line represents the group mean while the dotted lines represent the + Standard Deviation (SD). RD is right descent.

2.8: Knee moment profiles for the right leg of males during stair descent. Positive ‘y’ signifies an extension moment and negative ‘y’ is flexion. The solid line represents the group mean while the dotted lines represent the + Standard Deviation (SD). RD is right descent.

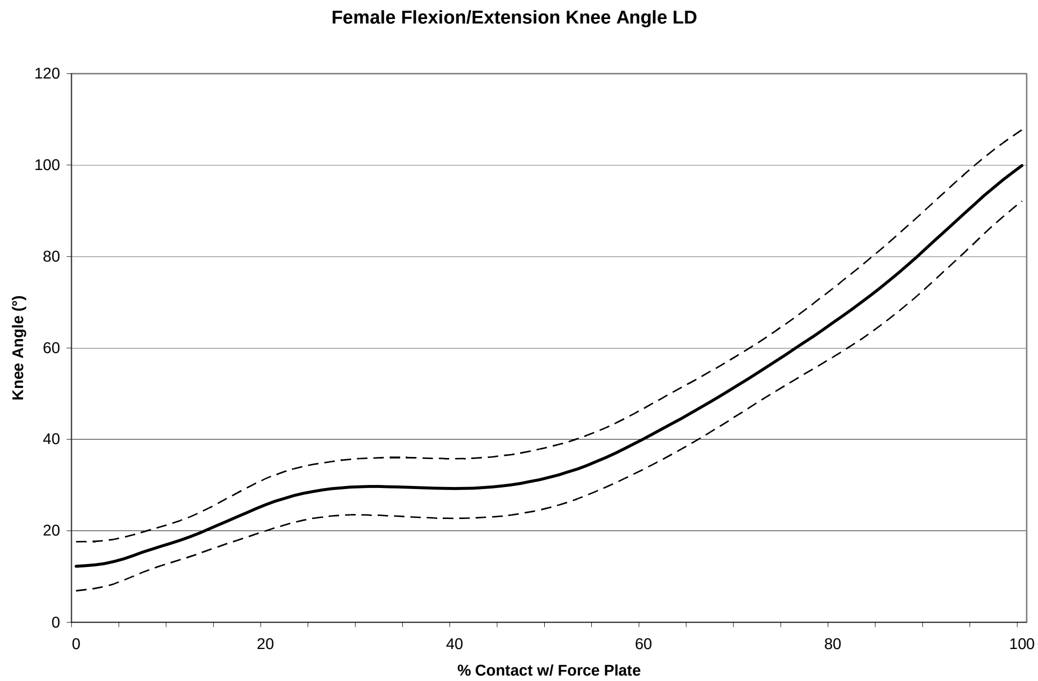
Appendix 3: Knee Angles of Males and Females during Stair Ascent and Descent



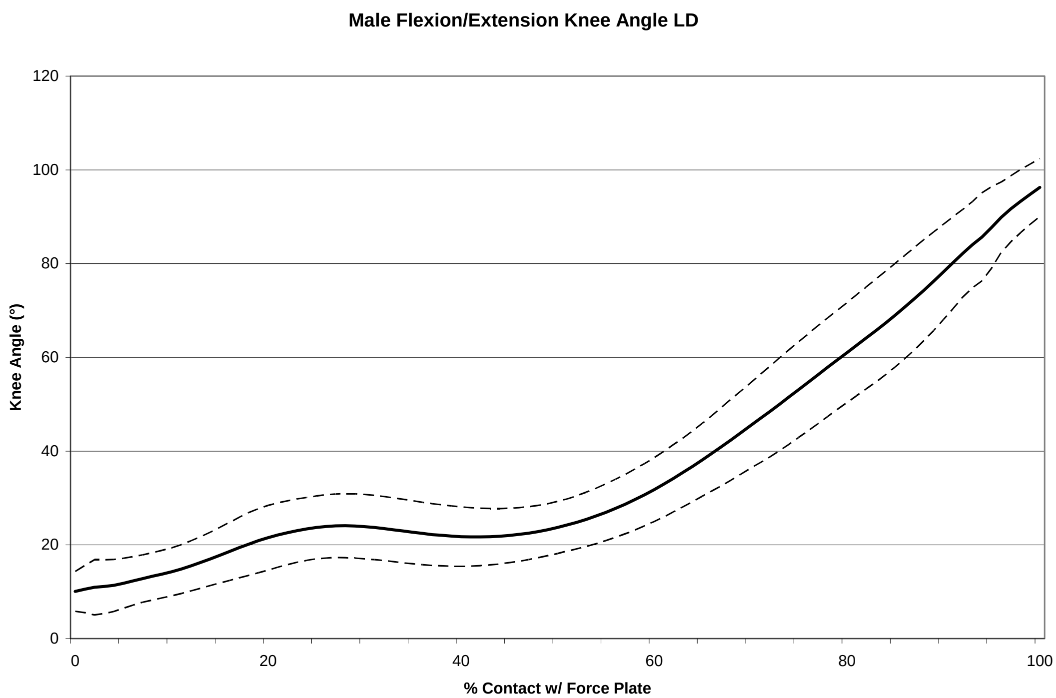
3.1: Knee angle profiles for the left leg of females during stair ascent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). LA is left ascent.



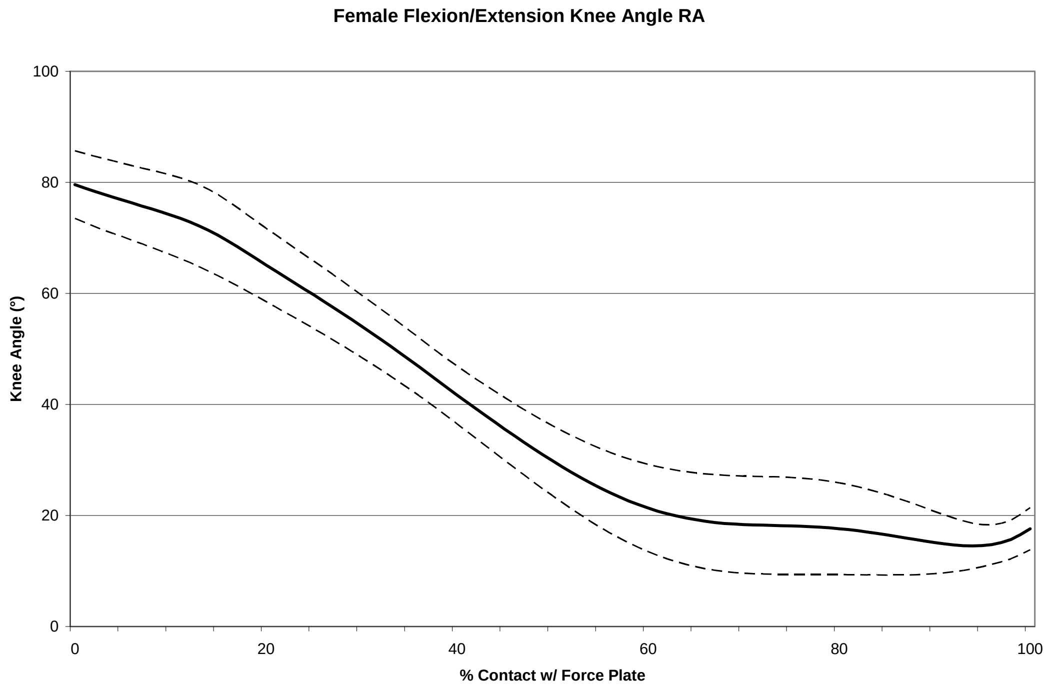
3.2: Knee angle profiles for the left leg of males during stair ascent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). LA is left ascent.



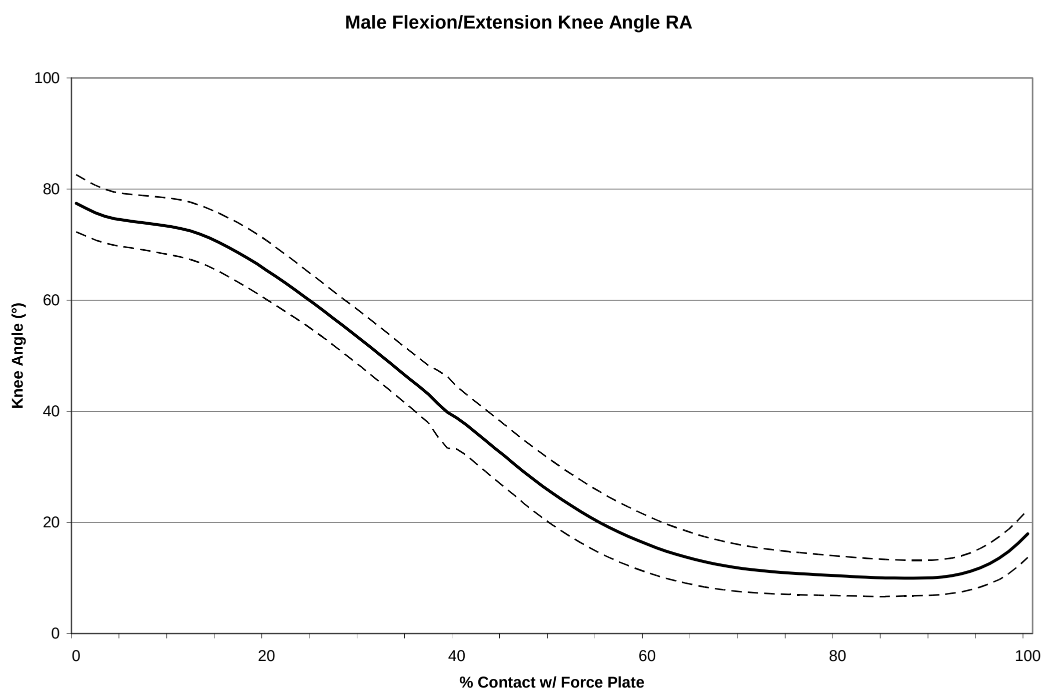
3.3: Knee angle profiles for the left leg of females during stair descent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the $\pm$ Standard Deviation (SD). LD is left descent.



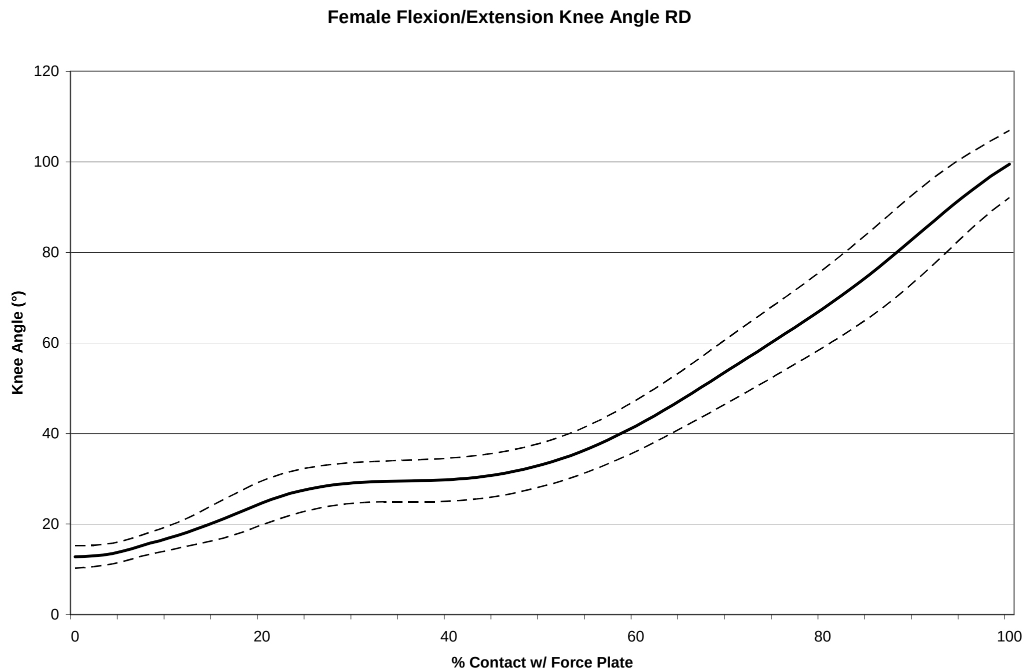
3.4: Knee angle profiles for the left leg of males during stair descent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the $\pm$ Standard Deviation (SD). LD is left descent.



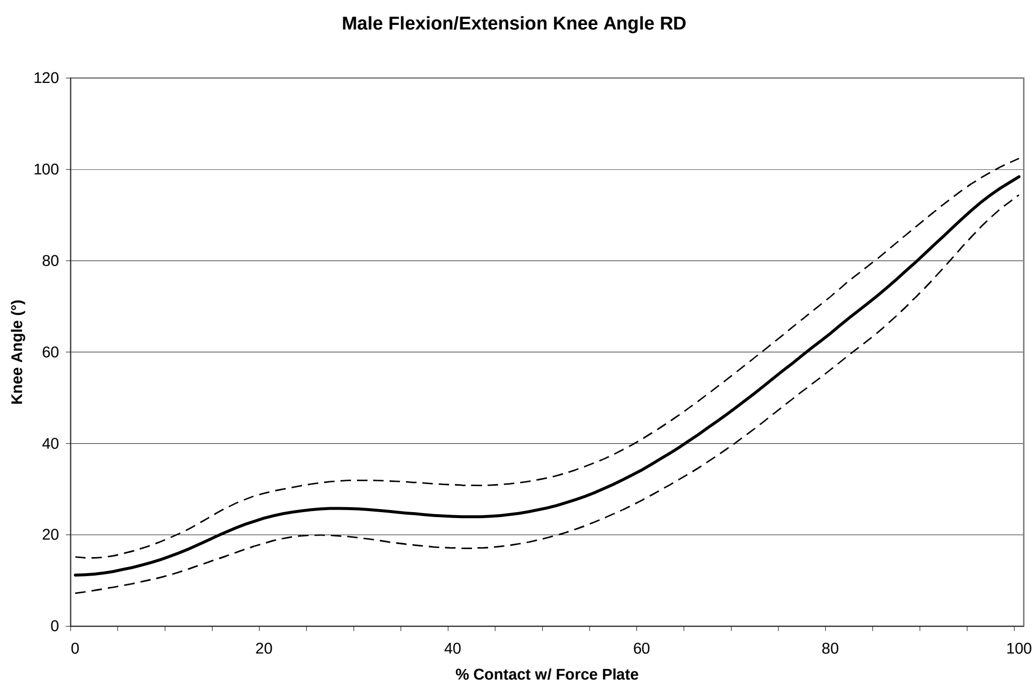
3.5: Knee angle profiles for the right leg of females during stair ascent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). RA is right ascent.



3.6: Knee angle profiles for the right leg of males during stair ascent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). RA is right ascent.

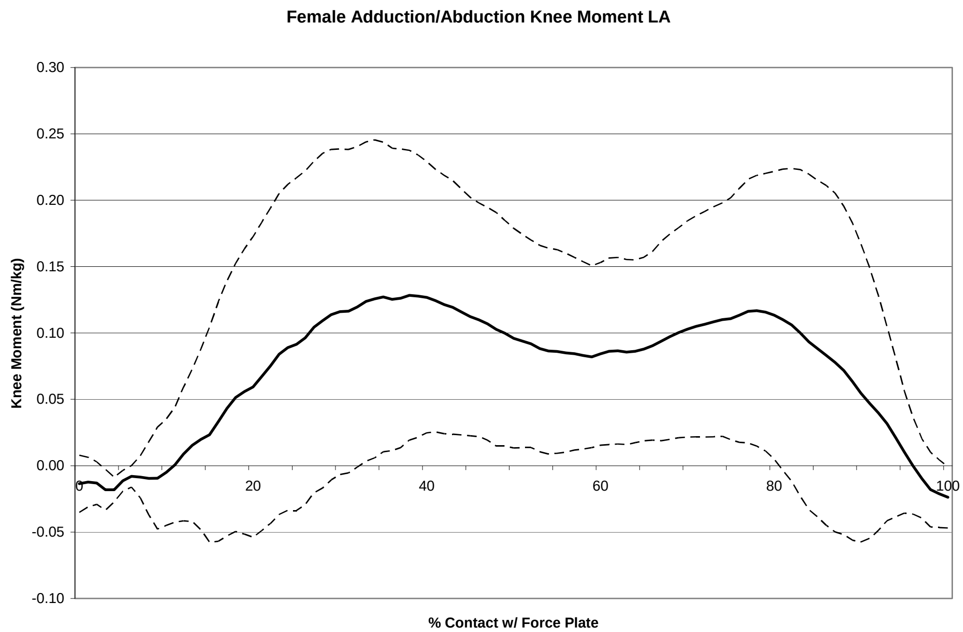


3.7: Knee angle profiles for the right leg of females during stair descent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). RD is right descent.

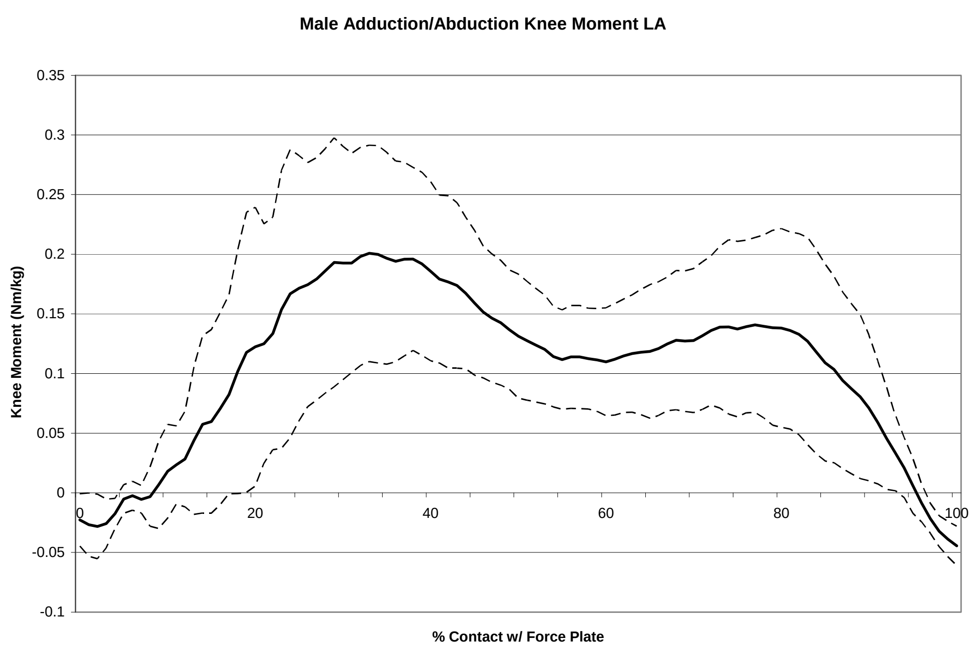


3.8: Knee angle profiles for the right leg of males during stair descent. 0° signifies full knee extension. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). RD is right descent.

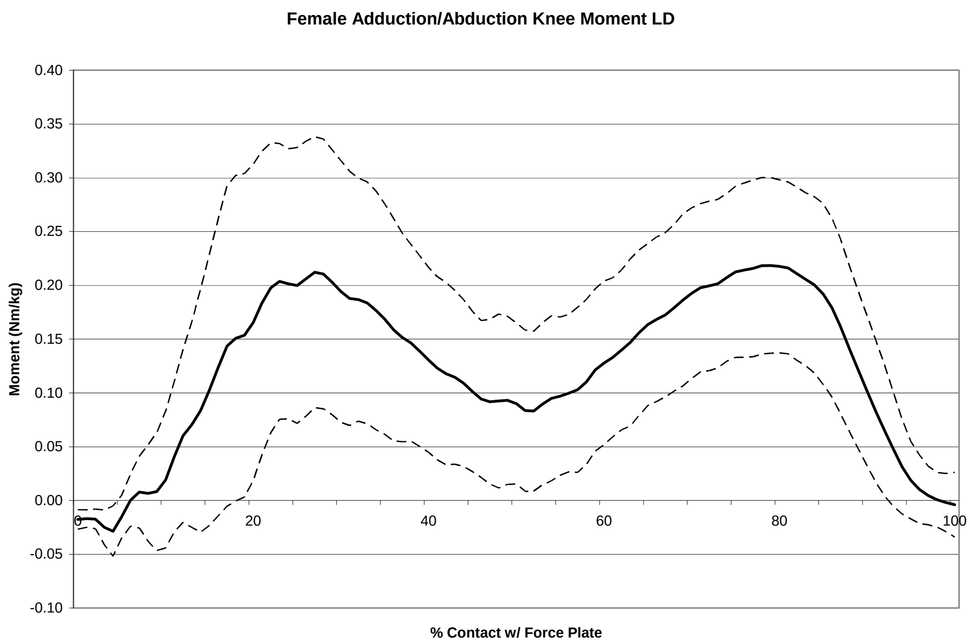
Appendix 4: Knee Adduction Moments of Males and Females during Stair Ascent and Descent



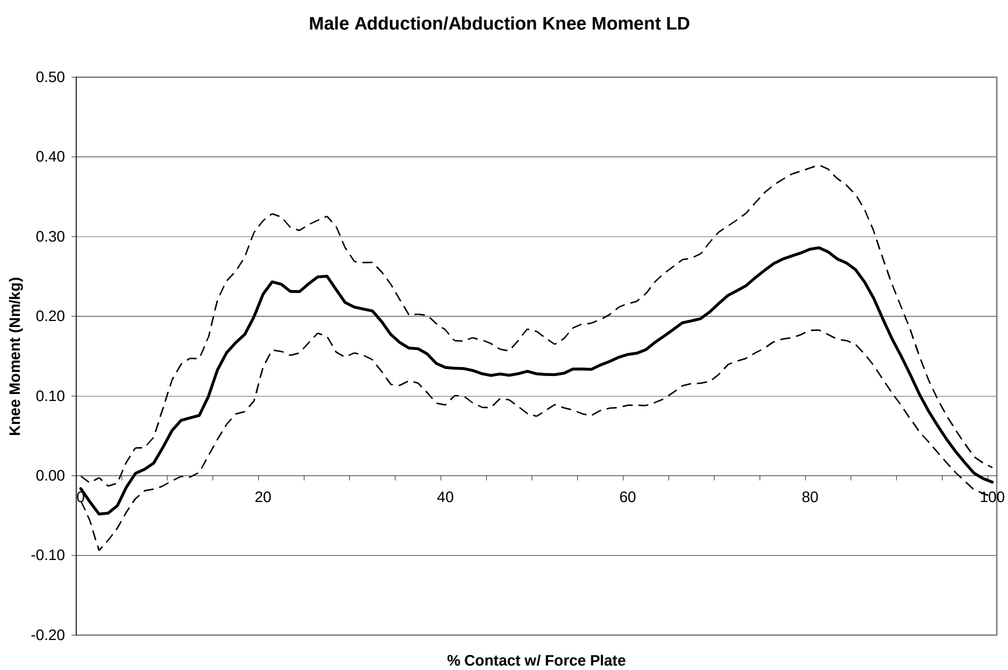
4.1: Knee adduction moment profiles for the left leg of females during stair ascent. Positive ‘y’ signifies an adduction moment and negative ‘y’ is abduction. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). LA is left ascent.



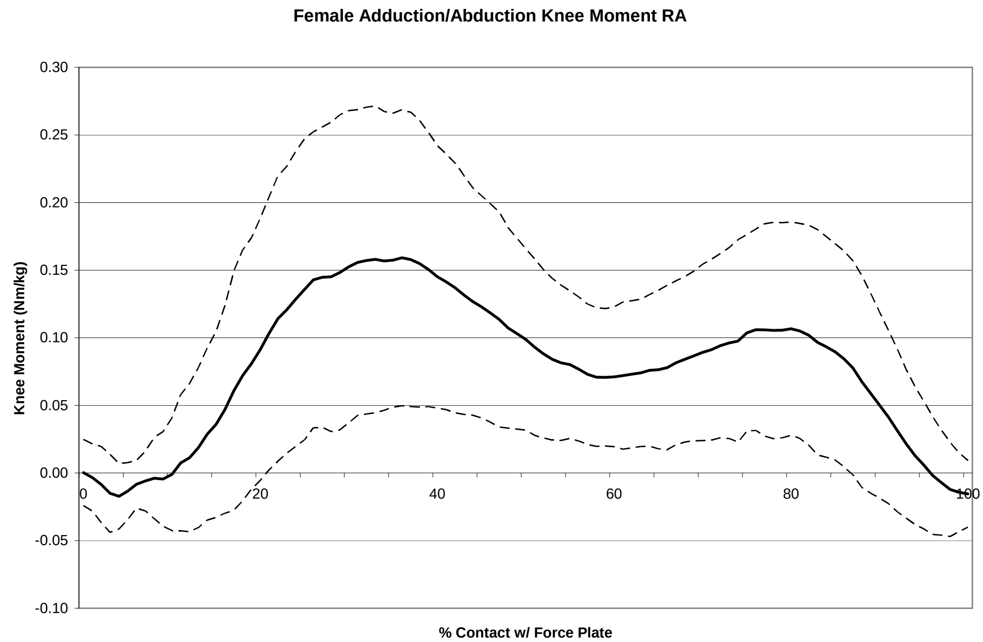
4.2: Knee adduction moment profiles for the left leg of males during stair ascent. Positive ‘y’ signifies an adduction moment and negative ‘y’ is abduction. The solid line represents the group mean while the dotted lines represent the ± Standard Deviation (SD). LA is left ascent.



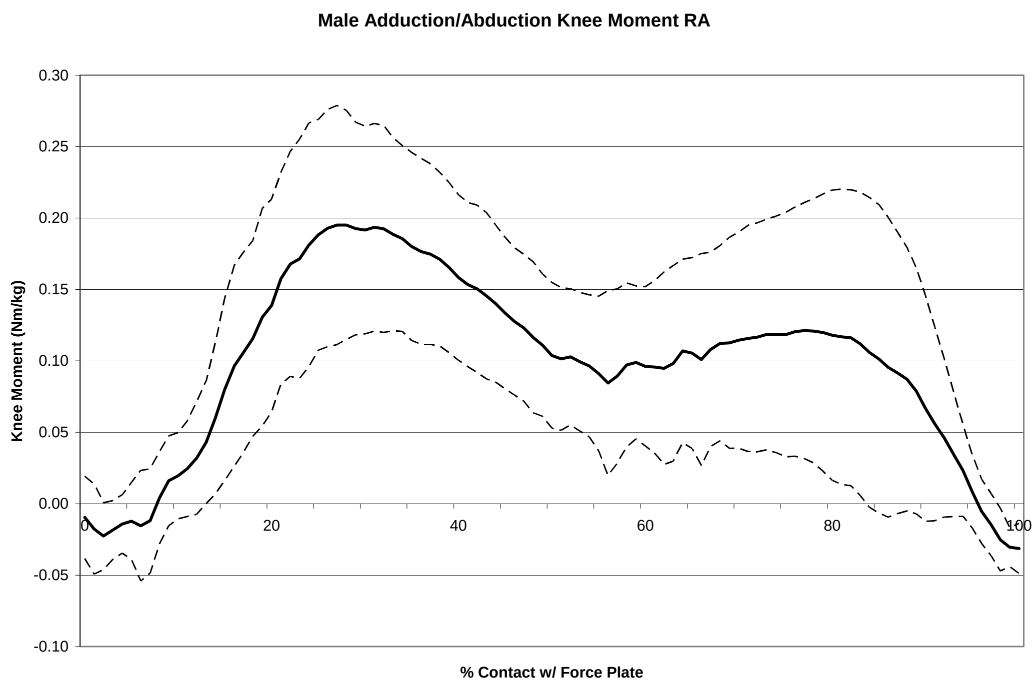
4.3: Knee adduction moment profiles for the left leg of females during stair descent. Positive ‘y’ signifies an adduction moment and negative ‘y’ is abduction. The solid line represents the group mean while the dotted lines represent the $\pm$ Standard Deviation (SD). LD is left descent.



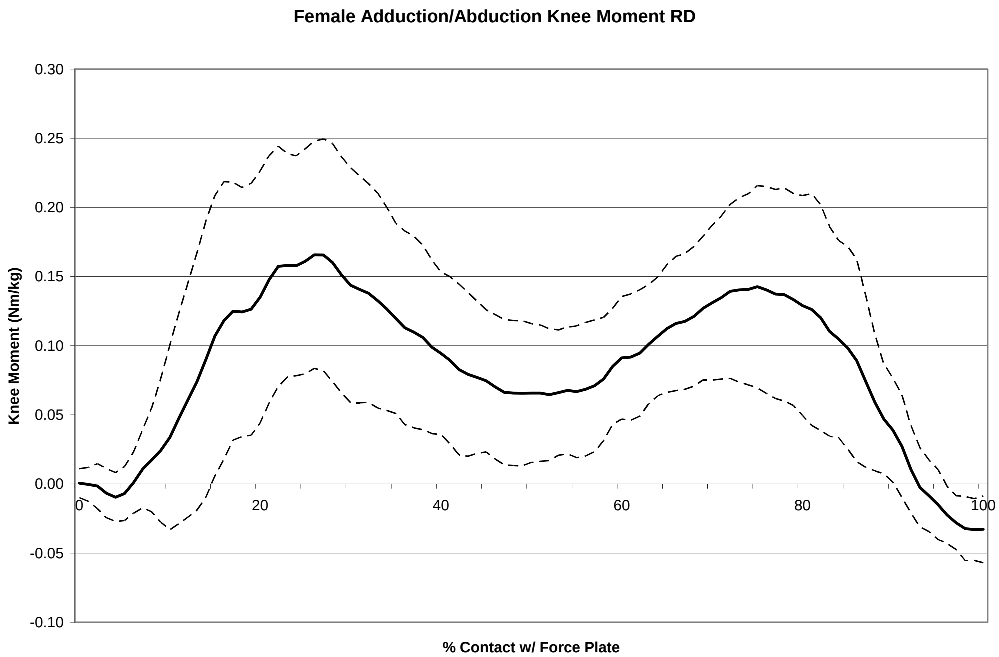
4.4: Knee adduction moment profiles for the left leg of males during stair descent. Positive ‘y’ signifies an adduction moment and negative ‘y’ is abduction. The solid line represents the group mean while the dotted lines represent the $\pm$ Standard Deviation (SD). LD is left descent.



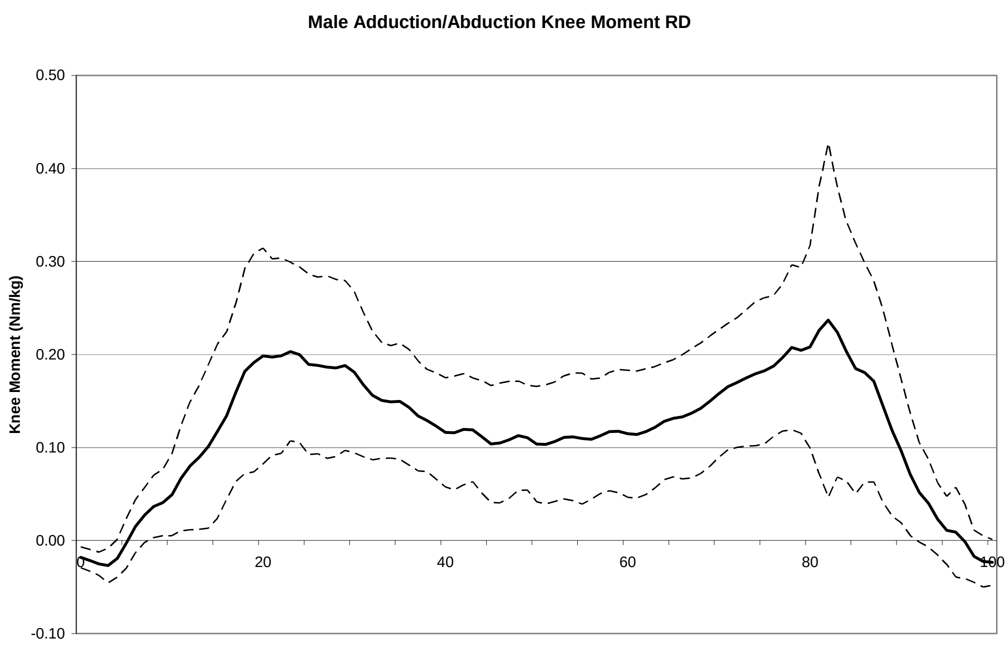
4.5: Knee adduction moment profiles for the right leg of females during stair ascent. Positive 'y' signifies an adduction moment and negative 'y' is abduction. The solid line represents the group mean while the dotted lines represent the + Standard Deviation (SD). RA is right ascent.



4.6: Knee adduction moment profiles for the right leg of males during stair ascent. Positive 'y' signifies an adduction moment and negative 'y' is abduction. The solid line represents the group mean while the dotted lines represent the + Standard Deviation (SD). RA is right ascent.



4.7: Knee adduction moment profiles for the right leg of females during stair descent. Positive 'y' signifies an adduction moment and negative 'y' is abduction. The solid line represents the group mean while the dotted lines represent the <u>+</u> Standard Deviation (SD). RD is right descent.



4.8: Knee adduction moment profiles for the right leg of males during stair descent. Positive 'y' signifies an adduction moment and negative 'y' is abduction. The solid line represents the group mean while the dotted lines represent the <u>+</u> Standard Deviation (SD). RD is right descent.

Appendix 5: Lower Limb Health Screening Form

<u>Lower Limb Health Screening Form</u>

Name: ____

Birth Date: ____ / ____ / _________ ***Gender:*** M F
D M Year

Questions:

1) Have you ever injured your knee badly enough to see a Doctor? Yes / No

If so, why? ____

2) When did this happen? ____

3) Have you had consistent pain in your knee, hip, or ankle in the last year? Yes / No

4) Have you ever had surgery on your knee, hip or ankle? Yes / No

If you answered yes, what was the surgery for and how long ago did the surgery take place?

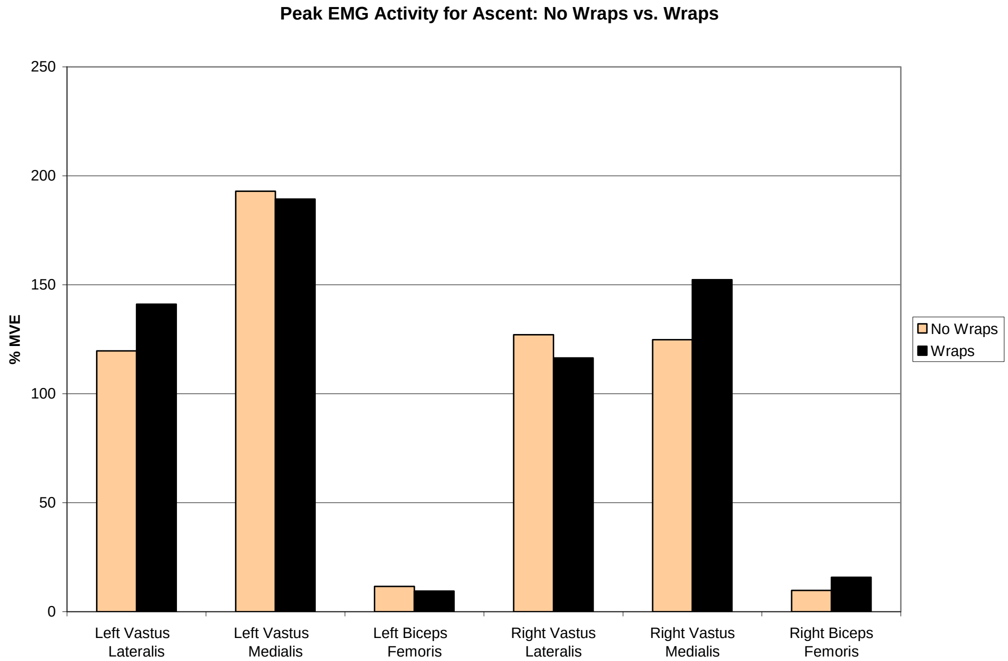
____

5) Have you been diagnosed with a condition that affects your knee, hip or ankle (i.e. Arthritis, bone diseases, etc.)? Yes / No

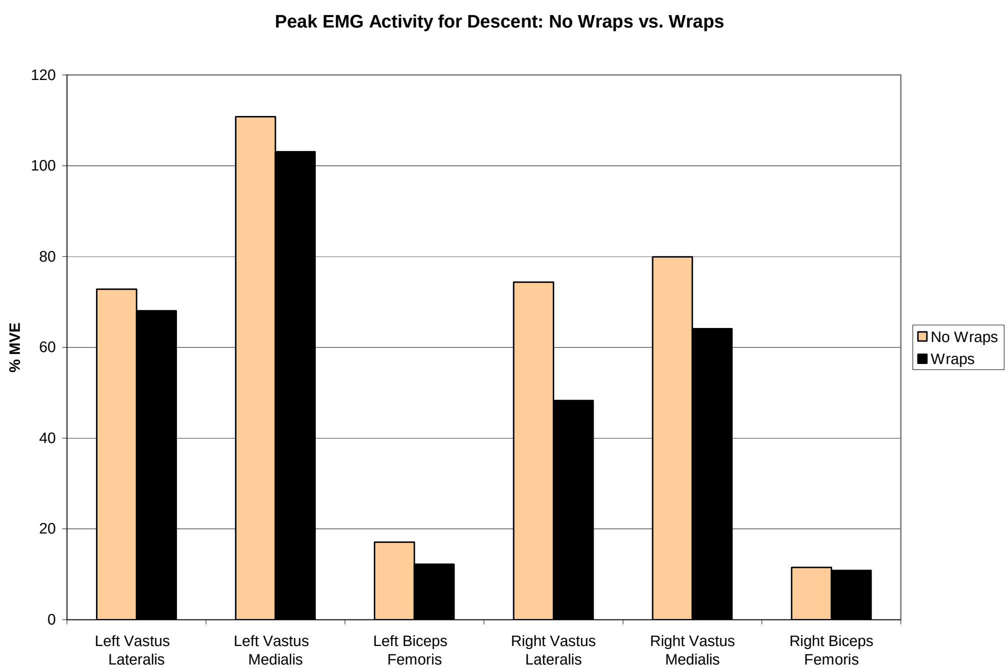
6) If yes, what is the condition? ____

Researcher Notes:

Appendix 6: Data from Electromyography (EMG) Collection with and without Thigh Wraps used for Tracking Plates



6.1: Peak electromyography (MVE) activity for the six muscles tested in Study II. Peak EMG is in units of % MVE. The trials were completed using the same MVE's for both conditions. This displays the result for stair ascent.



6.1: Peak electromyography (MVE) activity for the six muscles tested in Study II. Peak EMG is in units of % MVE. The trials were completed using the same MVE's for both conditions. This displays the result for stair descent.

Appendix 7: Statistical 'p' and 'F' Values for the Entire Data Set for Both Ascent and Descent

7.1: Statistical Values for Ascent Trials. First number represents the 'p > F' value. The number in brackets represents the 'F' value. Gender*Q-angle is the interaction. VL = Vastus Lateralis. VM = Vastus Medialis. BF = Biceps Femoris.

Ascent	p > F (F value)		
	Gender	Q-Angle	Gender*Q-Angle
VL Average EMG	0.0013 (12.12)	0.2115 (1.62)	0.9586 (0.00)
VM Average EMG	0.0033 (9.94)	0.6843 (0.17)	0.6043 (0.27)
BF Average EMG	0.8197 (0.05)	0.0297 (5.13)	0.7736 (0.08)
VL Occurrence of Peak EMG	0.0647 (3.63)	0.3426 (0.92)	0.0612 (3.73)
VM Occurrence of Peak EMG	0.8446 (0.04)	0.3892 (0.76)	0.1168 (2.58)
BF Occurrence of Peak EMG	0.8371 (0.04)	0.8775 (0.02)	0.9583 (0.00)
VL Peak EMG	0.0016 (11.69)	0.4840 (0.50)	0.4801 (0.51)
VM Peak EMG	0.0149 (6.53)	0.4960 (0.47)	0.8276 (0.05)
BF Peak EMG	0.3274 (0.99)	0.0531 (4.00)	0.1647 (2.01)
Peak Knee Extension Moment	0.5207 (0.42)	0.9052 (0.01)	0.0866 (3.10)
Occurrence of Peak Knee Extension Moment	0.5292 (0.40)	0.4779 (0.51)	0.1540 (2.12)
Peak Knee Abduction Moment	0.0963 (2.92)	0.5526 (0.36)	0.7548 (0.10)
Occurrence of Peak Knee Abduction Moment	0.1672 (1.99)	0.0410 (4.50)	0.5674 (0.33)

7.2: Statistical Values for Descent Trials. First number represents the ‘p > F’ value. The number in brackets represents the ‘F’ value. Gender*Q-angle is the interaction. VL = Vastus Lateralis. VM = Vastus Medialis. BF = Biceps Femoris.

Descent	p > F (F value)		
	Gender	Q-Angle	Gender*Q-Angle
VL Average EMG	0.001 (12.94)	0.3641 (0.84)	0.9505 (0.00)
VM Average EMG	0.0006 (14.03)	0.8165 (0.05)	0.2405 (1.42)
BF Average EMG	0.1447 (2.22)	0.3299 (0.98)	0.5339 (0.39)
VL Occurrence of Peak EMG	0.0007 (13.86)	0.9952 (0.00)	0.8409 (0.04)
VM Occurrence of Peak EMG	0.6675 (0.19)	0.4696 (0.53)	0.5104 (0.44)
BF Occurrence of Peak EMG	0.0009 (13.01)	0.0155 (6.46)	0.0707 (3.47)
VL Peak EMG	0.0002 (17.68)	0.5120 (0.44)	0.9553 (0.00)
VM Peak EMG	0.0116 (7.08)	0.9342 (0.01)	0.3555 (0.88)
BF Peak EMG	0.0466 (4.25)	0.9527 (0.00)	0.9619 (0.00)
Peak Knee Extension Moment	0.0521 (4.04)	0.2670 (1.27)	0.9063 (0.01)
Occurrence of Peak Knee Extension Moment	0.5949 (0.29)	0.4560 (0.57)	0.7453 (0.11)
Peak Knee Abduction Moment	0.0192 (6.01)	0.5452 (0.37)	0.2709 (1.25)
Occurrence of Peak Knee Abduction Moment	0.9730 (0.00)	0.2405 (1.42)	0.6978 (0.15)

Reference List

Kingma, I., deLooze, M. P., Toussaint, H. M., Klijnsma, H. G., Bruijnen, T. B. M., (1996). Validation of a full body 3-D dynamic linked segment model. Human Movement Science 15**,** 833-860.