Strategies and Adaptations Seen with
Unilateral Lower Limb Weighting during
Level Ground Walking and Obstacle Clearance Tasks

by

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Authors 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:

Previous lower limb weighting studies have placed a load on the legs bilaterally and tested different placement locations. It was previously determined that kinematic changes occur with greater masses and at joints proximal to weight placement [1]. Other studies have determined that these changes exist for a short adaptation period before parameters revert to a steady state [2]. Tasks that require voluntary gait modifications such as obstacle clearance have also been performed with lower leg bilateral weight addition [4]. In cases of normal obstacle clearance increased flexion at all three joints in the lower limb is needed to safely traverse the obstacle [3]. The goal of this study was to investigate joint kinematics and kinetics of unilaterally weighted participants using level ground force platform collection techniques, rather than a treadmill. It was hoped that this would allow for new insight into the adaptation periods and strategic motor pattern changes seen at the ankle, knee and hip.

Kinematic and force platform data were collected on two groups of 10 healthy male subjects. Group 1 (mean age = 23years, mean weight = 82.181kg, mean height = 1.798m) was a normal walking group and group 2 (mean age = 24.8years, mean weight = 79.901kg, mean height = 1.773m) was an obstacle clearance group. Both groups participated in 20 trials each of three different conditions; normal, weighted and weight off using a 2.27kg limb mass attached just proximal to the right maleoli markers. A repeated-measures two-way ANOVA was carried out on relevant variables in order to determine statistical significance.

Weight addition and removal affected the kinematics and kinetics of the normal walking and obstacle clearance groups. This effect was more prominent in the normal

walking group. If changes were seen, trials 1 through 3 were the locations showing a quick adaptation followed by a leveling off back to a new steady state in later trials.

Participants in the normal walking group chose to utilize the hip joint in order to control for weight addition and removal. Kinematically, changes in the hip joint angle occurred at all instances analyzed throughout the gait cycle with this effect being more prominent in the weight off condition. In conjunction with this, the hip joint energy generation increased during all phases of the gait cycle while the ankle and knee joints either decreased energy generation or increased energy absorption. In the obstacle group, participants also chose to increase flexion at the hip joint. However, the ankle joint also had either decreased plantarflexion or increased dorsiflexion at all the instances analyzed during the gait cycle. However, joint energy generation increases at these joints were only found during stance and at heel contact. The toe obstacle clearance values also showed a marked increase in trial 1 for the weighted condition which demonstrates a voluntary gait modification made by participants to safely traverse the obstacle that was quickly adapted for. Overall, the results found by previous studies using treadmill collection techniques were still seen in overground force platform data but they were not as robust.

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Table of Contents:

Authors Declaration ii

Abstract iii

Acknowledgements v

Table of Contents vi

List of Figures viii

List of Tables xi

Chapter 1
- 1.0: General Overview of Lower Limb Anatomy and Function 2

Chapter 2
Literature Review
- 2.0: Introduction 6
- 2.1: Studies Done Changing the Upper Limb Mechanical Properties 6
- 2.2: Limb and Gait Symmetries during Locomotion 10
- 2.3: Intersegmental Dynamics 11
- 2.4: Altering the Lower Limb Mechanical Properties 13
- 2.5: Obstacle Clearance and Avoidance Investigations 17
- 2.6: Research Questions 25
- 2.7: Research Hypotheses 26

Chapter 3
Experimental Methods
- 3.0: Subjects 29
- 3.1: Experimental Protocol 30
- 3.2: Measures of Analysis 33
- 3.3: Statistical Analysis 36

Chapter 4
Experimental Results
- 4.0: Group 1 – Normal Walking Group 41
- 4.1: Full Gait Cycle Graphs 42
- 4.2: Stance Peak Results 49
- 4.3: Toe Off Results 51
- 4.4: Swing Peak Results 55
- 4.5: Heel Contact Results 58
- 4.6: Minimum Toe Clearance 61
- 4.7: Group 2 – Obstacle Clearance Group 62

4.8: Full Gait Cycle Graphs 63
4.9: Stance Peak Results 70
4.10: Toe Off Results 71
4.11: Swing Peak Results 74
4.12: Heel Contact Results 75
4.13: Toe Obstacle Clearance 78
4.14: Obstacle Toe Clearance versus Adapted Obstacle Toe Clearance 79

Chapter 5
Discussion
5.0: Getting Started 82
5.1: Normal Walking Group 83
5.2: Obstacle Clearance Group 96
5.3: Normal Walking Group versus Obstacle Clearance Group 104
5.4: Obstacle Toe Clearance and Adapted Obstacle Toe Clearance 106
5.5: Results in the Real World and Future Research Directions 107

Chapter 6
Conclusions
6.0: Normal Walking Group 111
6.1: Obstacle Clearance Group 112
6.2: Global Conclusions 113

References 114

Appendices
Appendix A: Right Leg Dominance Questionnaire 117
Appendix B: Statistical Proof for Binning Trials 6-20 Together 118
Appendix C: Example of Post Hoc Tukey Tests Done 119
Appendix D: Post Hoc Tukey Test Results from Repeated Measures ANOVAs 120
i: Stance peak statistical results summary sheet 120
ii: Toe off statistical results summary sheet 120
iii: Swing peak statistical results summary sheet 121
iv: Heel contact statistical results summary sheet 121
v: Stance peak statistical results summary sheet (obstacle) 122
vi: Toe off statistical results summary sheet (obstacle) 122
vii: Swing peak statistical results summary sheet (obstacle) 123
viii: Heel contact statistical results summary sheet (obstacle) 123

List of Figures:

Figure 1: 52 Reflective Marker Set Up 30

Figure 2: 2.27kg limb weight and support belt 32

Figure 3: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint angles 43

Figure 4: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint angles 44

Figure 5: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint moments 45

Figure 6: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint moments 46

Figure 7: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint powers 47

Figure 8: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint powers 48

Figure 9: Hip joint peak angle during stance. This graph and the graphs like it to follow show the trial by condition interaction effects detected by statistical analysis. Each data point represents the average of 10 subjects for a particular discrete point analyzed in the gait cycle for a given variable (hip joint angle during stance in this case) with each particular experimental condition (normal, weighted and weight off) and trial (1 through 6) shown separately. Trial 6 represents trials 6 through 20 binned together and then averaged 49

Figure 10: Hip joint peak power during stance 50

Figure 11: Ankle joint angles at toe off 51

Figure 12: Hip joint angles at toe off 52

Figure 13: Ankle joint moments at toe off 53

Figure 14: Ankle joint powers at toe off 54

Figure 15: Toe velocity in the vertical direction at toe off 55

Figure 16: Knee joint peak angle during swing 56

Figure 17: Hip joint peak angle during swing 56

Figure 18: Hip joint angles at heel contact 58

Figure 19: Knee joint moments at heel contact 59

Figure 20: Hip joint moments at heel contact 59

Figure 21: Knee joint powers at heel contact 60

Figure 22: Minimum toe clearance 61

Figure 23: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint angles 64

Figure 24: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint angles 65

Figure 25: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint moments 66

Figure 26: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint moments 67

Figure 27: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint powers 68

Figure 28: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint powers 69

Figure 29: Ankle joint peak angle during stance (obstacle). This graph and the graphs like it to follow show the trial by condition interaction effects detected by statistical analysis. Each data point represents the average of 10 subjects for a particular discrete point analyzed in the gait cycle for a given variable (ankle joint angle during stance in this case) with each particular experimental condition (normal, weighted and weight off) and trial (1 through 6) shown separately. Trial 6 represents trials 6 through 20 binned and averaged together 70

Figure 30: Knee joint moments at toe off (obstacle) 72

Figure 31: Knee joint powers at toe off (obstacle) 73

Figure 32: Toe velocity in the vertical direction at toe off (obstacle) 74

Figure 33: Knee joint peak angle during swing (obstacle) 75

Figure 34: Ankle joint angles at heel contact (obstacle) 76

Figure 35: Hip joint angles at heel contact (obstacle) 76

Figure 36: Ankle joint powers at heel contact (obstacle) 77

Figure 37: Hip joint powers at heel contact (obstacle) 78

Figure 38: Toe obstacle clearance 79

Figure 39: Obstacle toe clearance versus adapted obstacle toe clearance 80

List of Tables:

Table 1: Average age, height and weight for participants in groups 1 and 2 ____________ 29

Table 2: Experimental groups and corresponding conditions ________________________ 31

Table 3: Average trunk girths and limb weight distances for both groups ______________ 33

Table 4: List of variables analyzed in each experimental group _____________________ 37

Table 5: List of post hoc Tukey test comparisons analyzed _________________________ 38

Table 6: Significant statistical results found between 2 experimental groups __________ 105

Table 7: Kinematic and kinetic results found in both experimental groups ____________ 106

Chapter 1

1.0: General Overview of Lower Limb Anatomy and Function

The ankle joint is made up of the distal ends of the tibia and fibula joining with the trochlea of the talus. It is reinforced laterally by the anterior talofibular, posterior talofibular and calcaneofibular ligaments and medially by the stronger fan like medial ligament. Although rotation, adduction and abduction are possible at the ankle, the most important movements are plantarflexion and dorsiflexion. During walking, the swing phase is initiated by plantarflexion at the ankle being produced by the muscular action of the gastrocnemius, soleus and plantaris muscles. The stance phase of walking is important because it supports the body through the swing phase and propels the body forward with push off force. Then, near the end of the swing phase, the ankle must dorsiflex in order for the foot to land properly. This action is accomplished largely by the tibilais anterior muscle along with other muscles in the anterior compartment of the leg (Moore et al, 1999). It is also important to realize that what muscles are doing at one joint, because of their gross anatomical position, cannot explain the complete role of those muscles. Zajac et al (1989) describes how the soleus muscle does indeed produce plantarflexion at the ankle but also can extend the knee. This effect at the knee is very prominent when the position of the knee is close to an extended position. The gastrocnemius muscle can also operate with the soleus muscle and produce the same actions as stated above but, as will be discussed later, this is not always the case (Zajac et al, 1989).

The knee joint consists of articulations between the lateral and medial condyles of the femur and tibia along with connections between the femur, tibia and patella as well. The knee is strengthened by ligaments both within the knee capsule and outside. Inside the knee capsule, the anterior and posterior cruciate ligaments cross each other connecting the femur and tibia. As well, the medial and lateral menisci are attached to the tibia to act mainly as shock absorbers and reduce the amount of friction between the femur and tibia during movement. Outside the knee capsule, the knee joint is strengthened further by five external ligaments; the patellar, lateral collateral, medial collateral, oblique popliteal and arcuate popliteal ligaments. During the swing phase of walking, the knee flexes due to the three hamstring muscles. However, during the stance phase of walking the knee must extend or resist excessive flexion due to body weight. This is accomplished by the quadriceps muscle group (Moore et al, 1999). Again, what is happening at the knee during gait cannot simply be described by the gross anatomy of the muscles acting there. Zajac (2002) shows that when the knee extensors are active in early stance, they actually decelerate the leg and cause the trunk to propel forward. Hence, the knee extensors main role in forward progression in early stance is actually to utilize eccentric contractions and transfer energy from the leg into the trunk (Zajac, 2002). As well, whereas at the ankle the gastrocnemius muscle can work together with the soleus muscle causing ankle and knee extension, at the knee the gastrocnemius muscle can operate with the biceps femoris muscle and actually cause flexion at both the knee and ankle joints (Zajac et al, 1989).

The body's most stable yet moveable joint is the hip. The hip is formed by the head of the femur fitting into the acetabulum of the pelvis forming a ball and socket joint. Contributing to hip joint stability are three large ligaments that reinforce the hip joint anteriorly, inferiorly and posteriorly named the iliofemoral, pubofemoral and ischiofemoral ligaments. Unlike the ankle and knee, the hip joint is multiaxial and has movements in flexion, extension, abduction, adduction, medial rotation, lateral rotation and circumduction. Like the knee, the hip also flexes during the swing phase of walking and extends during the stance phase. Flexion is accomplished by using primarily the iliopsoas muscle or iliacus/psoas major complex. Extension is achieved by the hamstrings, the adductor magnus and gluteus maximus muscles (Moore et al, 1999). Similar to the ankle and knee, gross anatomy cannot account for the entirety of the hip function during gait. Zajac (2002) uses the hip extensor muscles to demonstrate this. The hip extensors at early stance essentially perform the same duty as the knee extensors, to decelerate the leg and accelerate the trunk forward, but instead utilize concentric muscle contraction. Thus the knee and hip extensors act synergistically and illustrate that gross anatomy alone is not sufficient enough to explain the dynamics of locomotion (Zajac, 2002).

Chapter 2

Literature Review

2.0: Introduction

Under ideal and normal conditions of daily living, the body moves as one mechanically healthy system. The locomotor system is aware of the mechanical properties of all its parts and of the environment in which they act. Unfortunately, some individuals do not always have the luxury of operating under perfect conditions, forcing the body to move in mechanically compromising positions and in situations more complex than simple straight walking. Events within people's occupations occur where one side of their body may have altered mechanical properties. Such examples are within firefighting, military or even serving jobs where external loads are either worn or carried with the potential that only one side or limb of the body is affected. Some medical conditions such as stroke or Parkinson's disease usually only affect one side of the body, forcing the locomotor system to adjust to these changes and adapt to make the appropriate movements. Sport injuries like broken ankles and some types of industrial injuries resulting from unilateral pushing, pulling or repetitive movements may also affect only one limb, so altering limb mechanical properties with weight could become a form of rehabilitation. It is important to understand, both theoretically and practically, how a healthy human body copes with and adapts to a change in its normal mechanical system patterns.

2.1: Studies Done Changing the Upper Limb Mechanical Properties

Before adding weights to the lower limbs was studied, mechanical manipulations to the upper arms were commonly investigated due to the frequent use of the arms in volitional reaching. Donker et al (2002) tested walking situations where no mass was added and where mass was added to either both arms, the right arm or the right leg. All masses were 1.8kg and were attached to the wrists in the arm conditions and the ankle in the leg condition while walking occurred on a treadmill with speeds increasing from 0.5km/h to 5km/h. The goal of this study was to determine whether these added weight conditions affected the weighted limb individually or if the other non-weighted limbs showed changed locomotion patterns as well. It was found that when both arms were loaded, arm movement was decreased and muscle activity in the shoulders was increased. Interestingly though, when only the right arm was weighted, the contralateral arm increased its movement and shoulder muscle activity compared to normal. Similarly, when the right leg was weighted, the upper arms increased movement and muscle activity. Donker et al (2002) describe these results by suggesting that the motor command sent to muscles is always the same to both sides of the body. When only one limb is weighted but both limbs receive the same motor command, larger movements in the unweighted limb result. Additionally, a key finding was found when analyzing the arm movements across the various speeds of walking. At lower speeds of walking the arms swung at a 2:1 frequency ratio compared to the legs (Donker et al, 2002).

Hatzitaki and McKinley (2001) loaded one arm with 1kg and 2kg weights at the wrist when both arms were needed to reach a target within a certain time frame. In the weighted arm, it was found that the forearm and upper arm segment velocities decreased significantly compared to the unweighted arm, causing movement duration to be prolonged. These results were even more prominent by increasing the weight, suggesting that mass and reaching speed occurs with an inverse relationship. In order to accomplish the movement, subjects used earlier onset and prolonged bursts, shown through EMG analysis, rather than increasing the actual EMG activity used in the loaded arm. Like

Donker et al (2002), it was suggested that the motor system cannot send out a command to increase the amount of muscle activity in the weighted arm without doing the same in the unweighted arm. Bilateral motor signal projection is suggested (Hatzitaki et al, 2001).

Like the legs, Ghez and Sainburg (1995) showed that control of the arms also utilizes intersegmental dynamics. By testing control subjects against patients with deafferented nerves, they showed that reaching tasks involved the use of proprioception. In a normal coordinated reaching movement, the subjects had to move their arm forward by extending the elbow and flexing the shoulder angle followed by a quick reversal. Control subjects had no problem doing so because they could coordinate the joint timing, muscle activation levels and upper arm torques needed to produce a successful reaching pattern. However, the deafferented sensory patients, especially in the reversal phase of the reaching task, could not do so. This shows that proprioception plays a large role in controlling intersegmental dynamics during reaching tasks of the arm (Ghez and Sainburg, 1995).

Continuing with this research, Sainburg et al (1999), sought out to further examine the mechanisms involved with intersegmental dynamics in the arms in order to determine if the body uses anticipatory control and whether adaptation (learning) occurs. Healthy subjects had to trace a 20cm line with a forward and backward motion in less than 1.5s and with external forces being applied in either the medial or lateral direction. The order of these trials with loads were: 100 trials with the external medial load, 108 trials consisting of 100 medial loads and 8 surprise lateral loads and finally 100 external lateral load trials. It was found that in both the medial and lateral external load conditions, subjects were able to adapt to the external loads very quickly by making sharp accurate tracings of the line. However, within the surprise trials the initial direction errors, especially in the reverse (backward) portion of the trace, were ten times as great and resembled the trials already adapted for. This means that subjects were expecting what movement was going to be needed and that the nervous systems internal model of movement operates through anticipatory control (Sainburg et al, 1999).

Altering the actual mechanical properties of the limb is not the only way to affect the forces experienced by the arms. Lackner and Dizio (1994) altered the external environment in which people performed movements by placing them in a circular room rotating at 10rpm where velocity dependent Coriolis forces would be present. Subjects sat in the middle of the rotating room and had to reach to a visual target in three different conditions: prerotation, rotation at 10rpm and postrotation. Initially when the room was rotating, subjects were not accurate in their reaching trajectories because they missed the target in the direction opposite of rotation. However, within a few trials, subjects became accurate for the reaching tasks showing that the Coriolis forces were quickly adapted for. More interesting though is that when the room stopped rotating the subjects reaching accuracy was now again erroneous but in the direction opposite to when the room first started rotating. This tells us that yet again, subjects were using anticipatory control when reaching for the target because their movements were made thinking that the Coriolis forces would still be present. It is clear that the arms can rapidly adapt to changes in limb mechanical properties or the external environment which they act through anticipatory control, proprioception and continuous monitoring of motor commands (Lackner and Dizio, 1994).

2.2: Limb and Gait Symmetries during Locomotion

Much debate has focused on whether human gait is bilaterally symmetrical. Frequently, gait symmetry is assumed and many biomechanical studies are performed unilaterally to decrease set up time, number of markers and data evaluation. However, a review conducted by Sadeghi et al (2000) on studies investigating limb symmetrical patterns does not agree with the assumption that human locomotion is entirely symmetrical. In fact, people with no physical or neurological disabilities seem to have some degree of asymmetry between the two legs. It is suspected that these asymmetries are due to limb dominance and laterality in the brain. That is, a person's dominant leg plays more of a propulsion role in gait acting towards a specific goal while the other leg is used more for support or control during the dominant legs action (Sadeghi et al, 2000).

To examine this proposition, Sadeghi (2003) examined the ankle, knee and hip extensor/flexor muscle activity in both legs of normal people without disability. It was determined that viewing human gait with a global approach, both whole limbs together, versus a local approach, looking at each joint of each leg separately, produced different results. Globally, when considering function of each leg as a whole (all joints simultaneously), gait is quite symmetrical and no significant differences are seen. Conversely, if local comparison between each joint of each leg is made, statistically significant differences are seen at the ankle, knee and hip. Like above, these differences seem to be in the function of each leg, one is for propulsion while the other is for balance and coordination (Sadeghi, 2003).

Using force plates, along with kinematic data, many biomechanical variables can be calculated. The ground reaction force obtained from the force plates can itself be used to analyze gait symmetry. Giakas and Baltzopolous (1997) used time and frequency domain parameters of the ground reaction forces to examine gait symmetry in the stance phase of human locomotion. Within the time domain, the vertical and anterior-posterior directions showed symmetry between the left and right ground reaction forces but the medial-lateral direction did not agree with the symmetry assumption. On the other hand, within the frequency domain, harmonic analysis of the ground reaction forces revealed no significant asymmetries in all three directions. Overall, the authors concluded that human gait can be assumed symmetrical when looking at parameters in the sagittal plane because both time and frequency domain analysis produced symmetrical data in the vertical and anterior-posterior directions (Giakas and Baltzopolous, 1997). However, it should be noted that this study used a global (whole limb), rather than local (each joint separately) approach. As a result, the type of analysis being performed may dictate whether the assumption of sagittal plane symmetry may hold.

2.3: Intersegmental Dynamics

In order for the body to produce coordinated locomotion, the central nervous system (CNS) must have a very detailed awareness of the lower limbs. This not only includes the muscles that act at each joint but also the inertial properties of each connected segment of the lower limb, the passive effects exchanged between interconnected segments and the effect that gravity plays on them. Hoy and Zernicke (1985, 1986) showed how all these different parameters interact at the hip, knee and ankle joints during the swing phase of three different types of cat locomotion, a pace-like walk, a trot-like walk and a gallop. It was found that the muscle moments at each joint counteracted the inertial effects present. For example, at paw-off the ankle flexor muscle moment overpowered the inertial effect that would cause extension and at paw contact the opposite effect occurred. Interestingly, across all gait speeds these muscle moments also seemed to be in phase with the distal joint motion direction while out of phase with the proximal joint motion direction. Another key finding of this study was that the ankle and knee moments during the swing phase did not double in magnitude as the speed of locomotion doubled from walking to galloping. What this tells us is that intersegmental dynamics can control for type and speed of gait by utilizing swing phase kinetics to adjust muscle demands appropriately through changes in gravitational or inertial effects (Hoy and Zernicke, 1985, 1986).

Eng et al (1997) have also showed that intersegmental dynamics play an important role in locomotor situations where reactive control was needed. By perturbing the swing phase of human gait at two different points - at early swing and at late swing - two different reactive strategies were found. First, when early swing phase was interrupted, an "elevating strategy" was seen where subjects increased flexion at the hip, knee and ankle joints of the swing limb. This allowed for a greater toe clearance and an elevation of the body's COM. However, only the knee joint was controlled actively, meaning that the hip and ankle joints were passively coupled to produce the desired motion. The second strategy found during the late swing perturbation condition was called the "lowering strategy". It consisted of a flexed knee and a plantarflexed ankle that caused the swing leg to rapidly lower to the ground. In this case the knee and hip were actively controlled but still the ankle joint was passively caused to contact the ground sooner than normal due to the usual dorsiflexion at heel contact being absent. Thus, even though active control may have been used at some of the joints in the swing limb to properly clear unexpected perturbations, the CNS took advantage of passive intersegmental dynamics to help produce safe limb trajectories. It was speculated that these passive intralimb dynamics may be used to limit the number of different motor patterns needed to be produced by the lower limb (Eng et al, 1997).

Given the results, it was important to know whether these intersegmental dynamics are exploited by different ages of healthy people within a population. Contrary to her initial hypothesis, Ganley et al (2006) showed that the intersegmental dynamics during the swing phase of gait in 7 year old children and healthy adults were quite similar. It was actually expected that this would not be the case, but instead the hypothesis was proven wrong showing that by the age of 7 years, enough locomotion experience has been attained to produce patterns of intersegmental dynamics resembling adults. Different from previous studies, it was also shown that the knee joint during the swing phase relies heavily on active muscle activity while the ankle and hip move through passively controlled intersegmental dynamics (Ganley et al, 2006).

2.4: Altering the Lower Limb Mechanical Properties

Clearly, limb mechanics can be changed by altering the variables of segment mass, moment of inertia and length within the mechanical system. Two of the most prevalent ways this has been done was by using different weights and by changing the location that these weights were attached on the limbs. Hence, mass and moment of inertia are the two key variables (Royer et al, 2005). Four separate conditions were used that tested no load, a baseline measure that increased the total mass of the leg by 5% with the addition of weight over a natural distribution on the leg and smaller magnitudes of inertia, an increased mass condition (all weight attached to the proximal shank) and an increased moment of inertia condition (redistributed all the mass distally on the shank). It was found that both mass and moment of inertia variables significantly altered lower limb walking mechanics compared to the no load condition at the hip (Royer et al, 2005). In fact, both mechanical and metabolic power increased in the test conditions but with no real difference between the mass or moment of inertia conditions, suggesting that these two variables equally change the limb mechanics used at the hip (Royer et al, 2005). The interplay between mass and moment of inertia was still uncertain but one statistical difference found was that in the increased moment of inertia conditions, the participants swing time and stride time was increased (Royer et al, 2005).

Another study done by Martin et al (1990) looking at running and the swing leg, but using different limb weights at the thigh and foot, examined this interplay between mass and moment of inertia. Mass was manipulated by using 0.25kg or 0.50kg and moment of inertia by placing those masses on the thigh segment or foot segment. As expected it, was found that increased masses and more distally placed loads affected the limb kinematics and kinetics most substantially (Martin et al, 1990). For example, with respect to mass, the heavier 0.5kg masses placed at the foot and thigh caused the knee and hip moments, powers and instantaneous velocities to all increase significantly. However, with respect to moment of inertia, when the masses were placed at the thigh, the knee mechanics did not change and the hip mechanics changed only slightly (Martin et al, 1990). Collectively, increasing mass and placing it more distally affects the magnitude of the joint moments and powers most significantly. In this case, joints proximal to weight placement were affected. Since the net joint moments were small at the ankle and knee when the weight was placed on the thigh segment, this work determined that the hip was what generates and controls the motion of the swing leg segments by increasing muscular output strategically depending on where the mass was added. The hip accomplishes this by supplying adequate energy into the lower extremities during early swing so that when other segments demand increased energy output; it can be provided (Martin et al, 1990).

Furthermore, an important factor to consider was how long these effects of mass and moment of inertia affect walking patterns and whether or not these affects become accommodated. In a different study, participants were required to a have a 1.95kg weight attached to their distal leg over an 8 day period (only removed when sleeping or showering). It was found that immediate changes in gait occurred when first attaching the limb weight and, like previously seen above, these increases in net joint moments occurred at the more proximal knee and hip locations of the lower limbs (Smith et al, 2007). These changes were more prominent at the knee and were due to the weighted limb having a longer swing phase and the unweighted limb having a longer stance phase (Smith et al, 2007). However, it was determined that after the first 5 minutes of weight addition, no further changes were seen and they were maintained throughout the rest of the 8 day protocol until the weight was removed. At this time the net knee and hip joint moments returned back to their normal baseline values within 5 minutes again (Smith et al, 2007). This shows that the human body can adapt to altered lower limb mechanics very quickly if needed without having any long term adaptation to the additional weight and moment of inertia when it was again removed (Smith et al, 2007).

In relation to this type of work, is the body’s internal model or self representation of its mechanical parameters and appropriate movement patterns that is monitored continually in real time. Other research has looked at this model in order to determine if adding limb weight of 2kg around the shank segment center of mass (COM) may force this internal model to recalibrate itself and adapt to its new parameters of mass and moment of inertia. Using three conditions of treadmill walking; a pre, weight on shank COM and post condition, it was determined that ankle and knee joint kinetics changed abruptly after adding or removing the limb weight (Noble and Prentice, 2006). These results lasted approximately 25 strides in both situations. These results may be explained by the swing phase being altered the most and the steps taken with the unweighted limb being longer when compared to the pre condition (Noble and Prentice, 2006). One more key finding that this study showed was that during these first 25 strides or so when the weight was attached, dorsiflexion at the ankle occurred at earlier onsets, which may be showing a type of safety precaution taken by the participants in order to ensure appropriate toe clearance (Noble and Prentice, 2006). So, in both the above studies, it has been shown that adding and removing the limb weight causes immediate changes that plateau over time (Noble and Prentice, 2006; Smith et al, 2007). In any case it was clear that the internal model of motor control uses an adaptation period in order to update itself to new limb mechanical parameters. It does this because with added weight, the moment of inertia affecting the limbs becomes more pronounced forcing muscular activity and limb trajectory to compensate (Noble and Prentice, 2006).

In the current study, 2.27kg is only being added to one of the limbs, but Reid and Prentice (2001) added weight to both of the limbs at symmetrical locations. Participants walked over obstacles of low (4cm) to high (30cm) heights with no added weight and with 4.5kg of weight added at the center of mass of both legs. Again, here it was found that the flexion at all three joints (hip, knee and ankle) increased, but when the additional weight was attached, the most significant increase in flexion occurred at the knee (Reid and Prentice, 2001). Additionally, it was determined that as the obstacle heights increased, the knee joint changed toward an energy absorbing system whereas the hip joint increased its energy output, changing the leg mechanics from a defined step over the obstacle to more of a distinct propelling swing over the obstacle (Reid and Prentice, 2001). This knee-hip trade off did, however, produce equally efficient power profiles as in the trials with no weight added. So it seems that, again, the body can change the strategy used for obstacle clearance by utilizing intersegmental dynamics (Reid and Prentice, 2001).

2.5: Obstacle Clearance and Avoidance Investigations

The human body controls limb trajectory over obstacles under normal conditions using both active (muscle) and passive forces. Patla and Prentice (1995) had subjects' clear obstacles ranging in height from 0.5 to 30 cm. It was determined that the primary approach to clearing the obstacles was by flexing at all three of the lower limb joints; the hip, knee and ankle. However, within this data it was seen that rotational energy only increases at the knee with response to increasing obstacle heights, meaning that passive interaction between leg segments must be what was increasing the hip and ankle flexion angles, rather than extra muscle force (Patla and Prentice, 1995). Other evidence for this can be shown at the hip where with increasing obstacle height, the increased hip flexion was not associated with increased hip pull off power. Instead, it was modulated by controlling flexion at the knee and thus further showing evidence for the use of intersegmental dynamics (Patla and Prentice, 1995).

Many obstacle clearance investigations have focused on how changes in obstacle height effect the strategy used to safely clear the obstacle. Patla et al (1993) investigated the kinematic patterns of healthy subjects in both obstacle height and width manipulations. Obstacle heights and widths varied in size from small (6.7cm), medium (13.4cm) and large (26.8cm). They determined that subjects accurately perceived the different obstacle size manipulations and made appropriate movement adjustments to safely traverse the obstacles. In both the height and width conditions, toe obstacle clearance distances increased with increasing size with a 10cm difference between small and large obstacles. Simultaneously, toe velocity was actively slowed as a function of obstacle size increase as well, changing from 4.2m/s to 2.8m/s in the small and large conditions respectively. However, obstacle width did produce an increase of 2cm toe clearance in the small, medium and large conditions compared to obstacle height. It was suggested that extra safety precautions were taken with the obstacle width manipulations because the swing leg trajectory had to be maintained for a longer period of time. The strategies used to traverse the obstacles were determined to be accomplished by a combination of two techniques used unequally; the first being increased swing limb flexion accounting for 78% and the second being hip elevation (hip hiking) accounting for the other 22% of the combined strategy. The difference between obstacle height and width manipulation was that, in the obstacle width conditions, all three joints of the lower limb had increased flexion, but in the obstacle height conditions, only the hip and knee showed increased flexion (Patla et al, 1993).

Sometimes throughout the day locomotion does not always occur over solid terrain. MacLellan and Patla (2006) compared obstacle avoidance strategies between ground and a compliant surface travelling path using an obstacle of 30cm in height. They determined that, between the two different conditions step length, foot placement leading up to the obstacle and toe elevation, which was the distance from toe off to toe trajectory at obstacle clearance, did not vary. However, toe clearance, which was the vertical distance of the toe over the obstacle at the moment the toe crosses the obstacle, was significantly lower in the compliant surface condition. Still though, strategies involved with normal ground obstacle clearance tasks remained present when clearing the obstacles on the compliant surface. A knee strategy was used during the obstacle clearance phase in order to elevate the toe safely, while a hip strategy was used after obstacle clearance where energy was absorbed to decrease vertical foot velocity to help lower the limb safely at heel contact. On the whole, it was concluded that the compliant surface condition showed larger work profiles at all three joints compared to the ground condition and that toe clearance was not controlled for when stepping over obstacles. This was because toe clearance was smaller in the compliant surface condition, showing that the CNS did not include the depression of the compliant surface caused by weight-bearing into its movement strategy patterns. Instead, toe elevation was the variable controlled for because it was constant between both compliant and ground surface conditions (MacLellan and Patla, 2006).

Differences in these obstacle avoidance mechanics and strategies are not always the same between different age populations of people. Lu et al (2006) sought out to determine the alterations in obstacle clearance techniques between older and younger adults by using obstacle heights of 10, 20 and 30% of leg length. In the young adult group, the results found were not unexpected. With increasing obstacle heights, leading toe clearance and leading heel-obstacle distance was unchanged. In agreement with previous literature, Patla and Prentice (1995), this was accomplished through increased flexion at all three joints in the lower limb. Interestingly, the younger adults also chose to use a large amount of ankle eversion in their leading swing limb. However, in the older adult group, many biomechanical obstacle clearance variables were changed. The leading toe clearance and the trailing toe-obstacle distances increased with increasing obstacle height while the leading heel-obstacle distances shortened with increased obstacle height. This occurred because the older adult group chose to flex at their hip much more than any other joint of the lower limb and did not have this ankle eversion characteristic present at all like the younger adult group. Therefore, the older group used a “hip flexion strategy” that enabled them to safely and efficiently increase toe clearance for optimal obstacle clearance while the younger group used a “swing ankle eversion strategy”. It was thought that older adults choose to clear the obstacles in a manner that allows them to control the least amount of angular components. Such components include lead limb hip and knee crossing external rotations, stance limb crossing knee flexion and stance limb knee internal rotation. Since the hip was proximal and has the power to influence the rest of the leg, it was adopted to increase the safety precautions taken. Although increased toe clearance was a safety precaution, the shortened heel-obstacle distances seen in the older adult group may be interpreted as a risk of tripping. Nevertheless, it was suggested that hitting the obstacle with the heel or mid foot would result in a better manageable trip then if contacted with the toe, so the strategy used by the older adult group is still a safety precaution when clearing the obstacle (Lu et al, 2006).

Draganich et al (2004) were more interested in the trailing leg in obstacle clearance tasks. Again, younger and older adults were compared but with an added test condition of walking speed while clearing an obstacle 20cm in height. The walking speeds tested were a slow speed of less than 0.85m/s, a normal speed of 0.95-1.10m/s and a fast speed of greater than 1.20m/s. Speed was shown to have the most significant effect on the trailing limb in both the young and old adult groups. Hip flexion, knee flexion and ankle plantarflexion moments were all increased at early stance followed by hip extension and ankle dorsiflexion moments increasing in late stance. However, like the previous study explored above by Lu et al (2006), age also affected the trail limb differently when clearing the obstacle. The hip, knee and ankle adduction moments were all increased in the older adult group by 21% to 43%. Additionally, the angular velocity of hip flexion was 20% slower than that of the young adults. Together, Draganich et al (2004) suggest that, because older adults have decreased muscle capacity, movement is slowed in order to control the center of mass over the obstacle more precisely. Speed or age did not have an effect on the toe-obstacle distance of the trailing limb. With respect to the body's internal model of locomotion patterns that Noble et al (2005) discussed, the trail limb in obstacle clearance tasks seems to have a preset locomotor pattern that scales the appropriate joint flexions needed to safely maneuver over the obstacle. This locomotor pattern must be preset because the trail limb and obstacle are not within the subjects' field of view when crossing the obstacle (Draganich et al, 2004).

Lowrey et al (2007) performed a study similar to the ones above with the same type of purpose in trying to detect differences between young and old age populations, only two obstacles of 45% leg length separated by two step lengths were used instead.

Unlike the previous study, no age related differences were seen for the leading toe clearance variable. Another similarity between the young and old age groups was that within all obstacle obstructed trials, subjects decreased their step velocity while increasing their step time. However, overall the younger adults walked with step velocities that were significantly faster than the older adults while the older adults walked with a step width that was significantly smaller than the younger adults. In agreement with the previous study the older adults still had a notably decreased leading heel-obstacle distance at both obstacles. This is interesting because in the previous study, Lu et al (2006), an increased toe clearance with a decreased heel-obstacle distance was considered to be a safety precaution but clearly the results here do not agree. In order to more clearly determine and examine these discrepancies, Lowrey et al (2007) also calculated trunk roll (motion about the x-axis) and trunk pitch (motion about the z-axis) data. They found that both age groups across all obstacle conditions utilized a much larger range of trunk roll, especially during the steps between the two obstacles in the double obstacle conditions. In contrast, the older adult group had larger trunk pitch ranges (4.23°±2 compared to 3.60°±1) at the step prior to obstacle clearance but then returned to similar trunk pitch values as the younger adults when actually crossing the obstacles; again both age groups had increased trunk pitch. Thus the authors concluded that the older adults used an obstacle avoidance strategy that kept the obstacle more anterior in their step length by decreasing their heel-obstacle landing distance and walking with a decreased step velocity. By doing this, they are spending less time in single leg support. However, the older adult's step widths were also decreased making their base of support smaller. Even though a smaller base of support could potentially be risky, spending less time in single leg support over it was interpreted as a precautionary strategy. However, older and younger adults have similar trunk roll and pitch ranges during obstacle clearance which becomes worrisome when doing so over a smaller base of support. In doing this, hip musculature must be increased and in an older population this may not be possible due to neurological impairment or muscle degeneration being present (Lowrey et al, 2007).

The above studies have focused on young adults and the elderly population but Berard and Vallis (2006) looked at single and double obstacle clearance tasks between young adults and 7 year old children using obstacles of 20cm in height. Again step lengths, step widths, step velocities, toe clearances and take off distances were all collected, but an original distinct variable was also determined called the horizontal toe displacement at apex (HAD). It was defined as the distance between the toe and front edge of the obstacle horizontally at the instant when the foot is at peak height. First, differences were found in gait patterns between the age groups during normal flat ground walking: the children had longer step lengths, wider step widths and faster step velocities (after being normalized to leg height). Secondly, in the obstacle clearance trials, children chose to use strategies that incorporated larger toe clearances, smaller leading take off distances for the second obstacle, larger step widths and larger HAD values (meaning that children's lead foot peak trajectory was farther from the obstacle) compared to the adults. This type of strategy used by the children was described as a "squaring up" avoidance strategy by the authors. By stepping closer to the second obstacle (shortening the take off distance), the trail foot becomes closer to the lead foot while all along being within an increased base of support due to the increased step width, the children may have chosen a more cautious avoidance strategy. Together, the increased toe clearances and HAD values were also more cautious because they allow for the whole foot to land on the ground reducing the risk of a trip. Therefore, adults seem to possess the experience needed to avoid one or two obstacles safely because of their unchanging gait parameters between them. On the other hand, children must safely avoid the first obstacle in their path before the second obstacle can be planned for, and when it was planned for, a much more cautious approach was taken (Berard et al, 2006).

To this point, all of the studies reviewed have used subjects that, regardless of age, had fully intact knees. Byrne and Prentice (2003) tested right total knee arthroplasty (TKA) subjects against healthy subjects walking over obstacles of 6cm and 18cm in height with the replaced surgical limb as the lead limb over the obstacle. Surgical patients showed decreased knee flexor moments and positive knee flexor power during the elevation part of swing phase. These results were more pronounced in the larger, 18cm high obstacle condition. In addition, similar to previous obstacle clearance studies, Patla and Prentice (1995), the ankle, knee and hip flexion angles increased when crossing the obstacle for all people in the study. However, in the surgical group, the knee flexion was significantly decreased compared to healthy subjects. Interestingly, obstacle-toe clearance vertical distances were not different between the TKA and healthy subjects. The surgical TKA subjects accomplished adequate obstacle-toe clearance by swinging their lead toe laterally in the medial-lateral (frontal plane) over the obstacle and utilizing a slight amount of hip hiking as well. Byrne and Prentice (2003) described this strategy as hip internal rotation over the obstacle. Hip internal rotation used at the same time with knee flexion causes the lead swing toe to move laterally over the obstacle and achieve a safe toe clearance (Byrne and Prentice, 2003).

Throughout all these obstacle avoidance studies, it seems that people of all ages choose some strategy involving different joint activations, toe clearances, take off distances and landing placements depending on obstacle height and difficulty of the task. However, it was also evident that children of less than 7 years and older adults choose, although different, more safety cautious strategies in order to reduce the risk of a trip or fall.

2.6: Research Questions

The objective of this thesis research was to determine what strategies and movement patterns subjects utilize in order to adapt to unilateral changes in lower limb mechanical properties. Specifically, a 2.27kg weight was added just proximal to the ankle joint so that these issues could be looked at during situations of level ground walking and obstacle clearance tasks. Additionally, to date most of the lower limb weighting studies have been done using treadmills without force plate collection abilities. This research did not use a treadmill but level ground walking and force plates instead, which may have allowed for a greater insight into the kinetic changes that the lower limbs undergo when unilaterally weighted. Using a treadmill can be advantageous because multiple strides within the gait cycle can be collected simultaneously and continually monitored, but using force platforms allowed for the advantage of ground reaction forces being collected. Since ground reaction forces were obtained, joint powers and moments were calculated. Some of the individual research questions that are going to be addressed are:

1. Are the rapid changes consisting of an adaptation period to the new limb mechanics and the implementation of a new steady state pattern, shown in previous unilateral lower limb weighting studies, using treadmill collection techniques, also present when walking on level ground using force platforms in the stance and swing phase of the loaded limb?
2. How are obstacle clearance strategies changed to accommodate both the adaptation period and the new steady state pattern with the addition and removal of weight unilaterally? Specifically, is the margin of safety, shown through toe-obstacle clearance distance, maintained through voluntary actions of the dominant leg when it is weighted or is a whole new strategy seen?
3. Are the obstacle clearance strategies seen in the unweighted normal obstacle conditions still the same as those observed following limb weight adaptation when clearing the obstacle?

2.7: Research Hypotheses

It is hypothesized that analyzing the joint moments, joint powers and joint angles of the unilateral lower limb weighting data with the use of force platforms will allow for a more complete analysis of lower limb dynamics, giving insight into how the body adapts to changes in mechanical properties. Joint powers will be able to show us where significant changes in mechanical output are needed to accommodate the added mass. However, it is expected that the adaptation period to the limb weight will be similar to previous work (Noble and Prentice, 2006) where after approximately 25 strides of locomotion; patterns will stabilize to a steady state.

When clearing the obstacle with the 2.27kg mass attached just proximal to the ankle joint of the lower limb, the flexion at the hip and knee will most likely increase, causing the joint moments and powers acting there also to increase. These effects together will cause the obstacle-toe clearance distance to increase in magnitude making the margin of safety greater in the weighted conditions.

After the subjects have become adapted to the added mass, it is anticipated that the same obstacle clearance strategies used in unweighted normal obstacle trials will be observed. Since the body will have already compensated mechanically for the new limb properties, obstacle-toe clearance values should be similar to previous patterns, meaning the margin of safety will also be the same.

Chapter 3

Experimental Methods

3.0: Subjects

The research design used had two separate groups with 10 subjects recruited into each for a total of 20 subjects from the University of Waterloo student population. Complete ethics clearance was obtained for this research from the University Of Waterloo Office Of Research Ethics and each of the subjects became familiar with the procedures and inherent risks of the study before consenting to participation.

The subjects chosen were all of the male gender. In order to be a participant, the subjects were prescreened to fit certain height and weight requirements. Subjects were required to have a height ranging from 1.72m to 1.88m and a weight ranging from 78kg to 90kg. The average subject age, height and weight for groups 1 (normal walking) and 2 (obstacle group) are located in Table 1 below. In addition, the subjects needed to be right leg dominant and without any physical or neurological impairments that would affect normal locomotion. To ensure that subjects were right leg dominant a short standard questionnaire created to identify this variable was given before experimental set up (Appendix A).

Table 1: Average age, height and weight for participants in groups 1 and 2.

	Group 1	Group 2
(n) =	10	10
Age (yrs)	23	24.8
Height (m)	1.798	1.773
Weight (kg)	82.181	79.901

The reason for choosing subjects with similar mass was to ensure that the 2.27 kg being added to the lower limb was an approximately equal percentage of body mass and shank mass for all. Subjects with similar heights ensured that the obstacle height was approximately an equal percentage of leg length for all subjects. Right leg dominant participants were chosen because this is where the limb weight was added, it was the lead limb over the obstacle for obstacle clearance and due to its function in propulsion as stated by Sadeghi et al (2000, 2003).

3.1: Experimental Protocol

Subjects were first asked to change into their running shoes and a pair of shorts. Next, 52 reflective markers were attached bilaterally to the following anatomical landmarks: shoulder, iliac crest, ASIS, greater trochanter, head of the fibula, medial condyle, lateral malleolus, medial malleolus, medial heel, lateral heel, 1st metatarsal, 5th metatarsal, forefoot and great toe. The remaining 24 reflective markers consisted of six rigid plates, all with four markers each. These were located bilaterally on the thigh and shank segments with the final two plates being on the trunk and posterior pelvis segments (Figure 1).

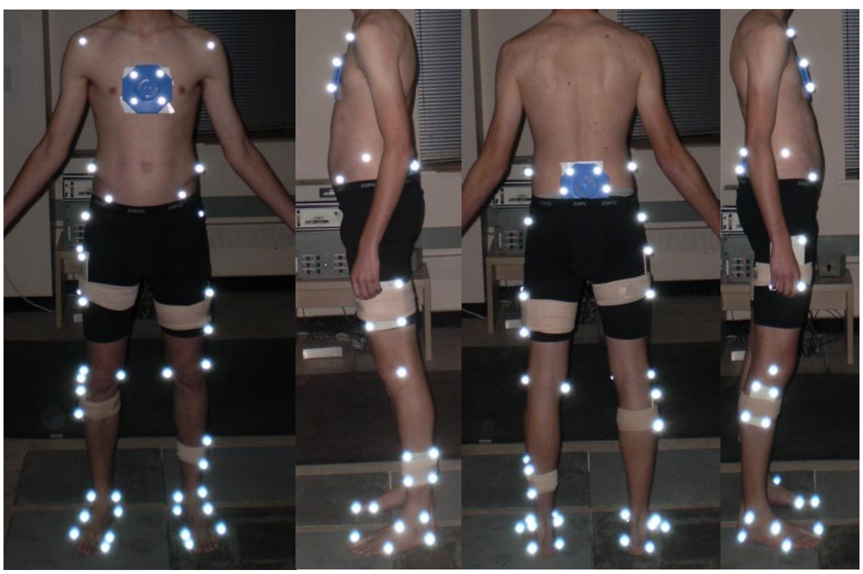


Figure 1: 52 Reflective marker set up

Kinematic tracking of these markers was done using 8 Vicon motion capture cameras (Vicon Motion Capture Systems, Los Angeles, CA) at a sampling rate of 64Hz. Placed in the middle of the 8 Vicon camera capture volume were two force platforms

(AMTI, Waterdown, MA) to capture ground reaction forces of the subjects over a full gait cycle. The force platform data was collected at a sampling rate of 576Hz.

Subjects were then randomly assigned to one of the following two experimental groups each participating in different tasks shown in Table 2 below.

Table 2: Experimental groups and corresponding conditions

Number of Trials	Group 1	Group 2
20	normal walking with no limb weight	obstacle clearance with no limb weight
20	normal walking with 2.27kg limb mass	obstacle clearance with 2.27kg limb mass
20	normal walking with no limb weight	obstacle clearance with no limb weight
20	normal walking with 2.27kg limb mass	
20	obstacle clearance with 2.27kg limb mass	

In all of the trials (group did not matter), subjects were required to step on the first force platform with their right foot or weighted foot, take another step with the unweighted limb onto the second force platform and then in the obstacle conditions step over the obstacle with the right or weighted limb. In all conditions the right leg was always the weighted limb and in respective trials the lead limb over the obstacle. The first group has an extra 40 trials within it so that the measure of adapted obstacle clearance with limb weight can be collected in order for proper comparison to the normal obstacle trials in group 2. Twenty trials were chosen for each condition because it has been shown previously (Noble and Prentice, 2006) that this is a sufficient amount of time for subjects to take enough strides to become adapted to the change in lower limb mechanical properties. Since data collection only occurred on the subjects while walking in one direction, a lab assistant rolled the subject back to the starting collection position in each of the weighted and weight removal trials. A computer chair was used to accomplish this because it provided a foot support rest for subjects weighted limb to rest upon in between trials in order to ensure that adaptations and changes seen did not occur in between collection trials. In all obstacle trial conditions, the height of the obstacle was 30cm and placed mid way through the subjects swing phase. An obstacle height of 30cm was chosen because it was a high percentage of subjects' leg length and has been previously shown to be a challenging task, forcing subjects to make adjustments at the ankle, knee and hip (Patla and Prentice, 1995; Berard et al, 2006). The obstacle height was not scaled to each participant because all of the subjects participating in the study were of relatively equal size and in the real world obstacles are the same for all people traversing them. The 2.27kg limb weight was chosen because it caused a large increase in the subjects' shank segment mass and was a large percentage of normal shank mass. Previous studies (Royer et al, 2005; Smith et al, 2007) have used similar limb weights of 1.95kg as well. Placement of the 2.27kg limb weight was just proximal to the ankle malleoli markers and supported by a belt that did not allow the weight to slip down the leg during trials while unaffecting movement at the same time (Figure 2).

Figure 2: 2.27kg limb weight and support belt

Lastly, specific measurements, other than height and weight, of the subjects were collected in order to create a Visual 3D model and perform calculations needed to obtain joint kinematics and kinetics. These further measurements were trunk girth (at sternum level) and the distance of the added limb weight from the knee. The average values for both of these variables in both of the experimental groups are located in Table 3 below.

Table 3: Average trunk girths and limb weight distances for both groups.

	Group 1	Group 2
Trunk Girth (m)	0.225	0.23
Limb Weight Distance from Knee (m)	0.323	0.322

3.2: Measures of Analysis

The three dimensional marker and force platform data was exported into Visual 3D software (C-Motion Systems, Inc.) in the C3D file format which contains both the analog force platform data and 3D kinematic marker data. Using this software the 3D marker data was filtered using a dual pass 4th order Butterworth, low pass filter with a cut off frequency of 6Hz. The force platform data was filtered using a dual pass 4th order Butterworth, low pass filter with a cut off frequency of 15Hz.

In Visual 3D, an anatomical rigidly linked model was created having feet, shanks, thighs, a pelvis and a trunk segment. The foot, shank and thigh segments were modeled as frusta's of cones while the pelvis and trunk segments were modeled as elliptical cylinders. Using Dempster's anthropometric tables (Winter, 2005), subject specific segmental mass, segment center of mass and segmental moment of inertia were calculated. Then, using Newtonian inverse dynamics (Winter, 2005), joint reaction forces and joint moments (normalized to body weight) were calculated.

Due to the weight being added to the right limb in certain trials, some of the standard calculations had to be modified in order to encompass the correct segment parameters. When calculating the centre of mass, the assumption that the ankle weight is a point mass embedded in the F_z (vertical) axis was made. So the centre of mass calculation in the z direction or longitudinal axis of the shank was the only modification needed and this therefore, produced a $COM_{zhybrid}$ calculation. It included the normal shank calculation plus the added weight component using the measurement taken earlier of the distance between the knee and ankle weight (Z_{wt}).

$$COM_{zhybrid}$$
$$= ((P_{shank}*Z_{cogshank}) + (P_{weight}*Z_{cogweight}))/ P_{total}$$
$$= (((P_{shank})*(Z_{prox} + r_{prox}*(z_{distal}-z_{prox}))) + (P_{weight}*Z_{wt}))/ P_{total}$$

where:
P_{shank} = segment mass proportion,
P_{weight} = weight mass proportion,
r_{prox} = segment COM/length,
Z_{wt} = distance from knee to the ankle weight (Robertson et al, 2004).

The moment of inertia calculations needed the same assumption as above. Then using the parallel axis theorem the segment moment of inertia about the hybrid shank COM was obtained using the centre of mass inertia plus the added inertia due to the point mass (I_{wt}).

$$I_{total}$$
$$= I_{shankcog} + I_{wtcog}$$
$$= (m(k_{cog}l)^2+md_{hybridCOM}{}^2) + m_{weight}d_{weight}{}^2$$

where:
k_{cog} = radius of gyration about the COM,
l = shank segment length,
$d_{hybridCOM}$ = distance from shank COM to $COM_{zhybrid}$,
d_{weight} = distance from $COM_{zhybrid}$ to point mass (Robertson et al, 2004).

In order to evaluate the changes seen in walking and obstacle clearance tasks caused by the added limb weight, a 2D analysis of the following variables were analyzed: joint angles, joint moments and joint powers. All three of these measures were examined at the ankle, knee and hip joints within the local axis system of each segment using Visual 3D software.

Since the adaptation periods to the changes in lower limb mechanics needed to be identified and for statistical purposes, discrete values of the above three variables were examined at different points within the gait cycle. In the normal walking trials, joint angles, joint moments and joint powers were looked at during peak stance, at toe off (TO), during peak swing and at heel contact (HC) of the weighted right limb. In the obstacle trials, the same points as above were also examined in the weighted right limb.

Additionally, four other variables were examined in order to further identify changes in limb mechanics and patterns. These variables were toe velocity in the horizontal direction (forward progression), toe velocity in the vertical direction (longitudinal axis), minimum toe clearance (min TC) and toe obstacle clearance (TC). The toe velocities were examined for both group 1 (normal walking) and group 2 (obstacle group) at the discrete points of toe off and heel contact. These toe velocities were calculated using Visual 3D software and the reflective marker placed on the right great toe of the foot. Minimum toe clearance was only calculated for group 1 in the normal walking trials. It was defined as the minimum distance between the great toe reflective marker on the right foot and the ground at the lowest point during the swing phase of the gait cycle. Toe obstacle clearance was only calculated in group 2 and for the last 20 trials in group 1 where obstacle clearance was a part of the task. Toe obstacle clearance values were determined by placing additional markers on the obstacle. The vertical distance between the great toe reflective marker and the obstacle markers at the point when the toe was directly over the obstacle was considered to be the toe clearance variable or a measure of the margin of safety (MOS). This margin of safety measure determined during the obstacle clearance trials was used to show the differences in safe clearance, if any, between weighted and un-weighted conditions.

3.3: Statistical Analysis

Analysis of all the variables discussed above was done using SAS Version 9.0 (Cary, NC) and the statistical tests of two-way repeated measures ANOVAs having the within factors of condition and trial. The factor of condition consisted of no weight, weighted and weight off for all the two-way repeated measures ANOVAs. The factor of trial consisted of trial numbers 1, 2, 3, 4, 5 and 6. The reason for only having 6 trials included in the trial factor was because of the 20 trials within each test condition, trials 6 through 20 were collapsed together. After performing statistical two-way repeated measures ANOVAs on these 15 trials to determine if any interaction effects were present between them, it was determined that collapsing trials 6 through 20 was acceptable. In fact, of the 41 variables analyzed in the normal walking group, only one variable showed a significant interaction effect. Of the 42 variables analyzed in the obstacle clearance group only two variables showed an interaction effect. Because of this, it was deemed both acceptable and necessary to collapse these 15 trials together. This allowed for a much stronger statistical analysis by decreasing the possibility of obtaining a significant result due to the high volume of data and chance alone. Table 4 below shows a list of the specific variables that were analyzed in both group 1 (normal walking) and group 2 (obstacle group) on the right limb while Appendix B shows the statistical evidence that allowed for trials 6 through 20 being collapsed together.

Table 4: List of variables analyzed in each experimental group

(1) Normal Walking Group			
Stance Max/Min	Toe Off	Swing Max/Min	Heel Strike
Ankle Angle	Ankle Angle	Ankle Angle	Ankle Angle
Ankle Moment	Ankle Moment	Ankle Moment	Ankle Moment
Ankle Power	Ankle Power	Ankle Power	Ankle Power
Knee Angle	Knee Angle	Knee Angle	Knee Angle
Knee Moment	Knee Moment	Knee Moment	Knee Moment
Knee Power	Knee Power	Knee Power	Knee Power
Hip Angle	Hip Angle	Hip Angle	Hip Angle
Hip Moment	Hip Moment	Hip Moment	Hip Moment
Hip Power	Hip Power	Hip Power	Hip Power
	Toe Vel. X		Toe Vel. X
	Toe Vel. Z		Toe Vel. Z
Minimum Toe Clearance			
(2) Obstacle Clearance Group			
Stance Max/Min	Toe Off	Swing Max/Min	Heel Strike
Ankle Angle	Ankle Angle	Ankle Angle	Ankle Angle
Ankle Moment	Ankle Moment	Ankle Moment	Ankle Moment
Ankle Power	Ankle Power	Ankle Power	Ankle Power
Knee Angle	Knee Angle	Knee Angle	Knee Angle
Knee Moment	Knee Moment	Knee Moment	Knee Moment
Knee Power	Knee Power	Knee Power	Knee Power
Hip Angle	Hip Angle	Hip Angle	Hip Angle
Hip Moment	Hip Moment	Hip Moment	Hip Moment
Hip Power	Hip Power	Hip Power	Hip Power
	Toe Vel. X		Toe Vel. X
	Toe Vel. Z		Toe Vel. Z
Toe Obstacle Clearance			

Once trials 6 through 20 were binned together, the two-way repeated measures ANOVAs were conducted on the three conditions (normal, weighted and weight off) and six trials (1 – 6). If they revealed a condition by trial interaction effect with an alpha significance level of 0.05 or less, post hoc Tukey tests were conducted on the least squares means of the condition by trial interaction effect to determine where significant changes in specific variables occurred. Again these differences were only deemed significant if an alpha significance level of 0.05 or less was present. There were 18 specific post hoc Tukey tests performed for each variable listed above and these specific patterns that were looked for are listed in Table 5 below. See Appendix C as well for an example of how these post hoc Tukey tests were performed.

Table 5: List of post hoc Tukey test comparisons analyzed

Post Hoc Tukey Test Patterns	
Trial 1	normal vs. weight
	normal vs. weight off
	weight vs. weight off
Trial 2	normal vs. weight
	normal vs. weight off
	weight vs. weight off
Trial 3	normal vs. weight
	normal vs. weight off
	weight vs. weight off
Trial 4	normal vs. weight
	normal vs. weight off
	weight vs. weight off
Trial 5	normal vs. weight
	normal vs. weight off
	weight vs. weight off
Trial 6	normal vs. weight
	normal vs. weight off
	weight vs. weight off

If these two-way repeated measures ANOVAs did not reveal a condition by trial interaction effect ($\alpha > 0.05$), then main effects due to condition were looked at. For a main effect to be present, the subject by condition effect had to have an alpha significance level of 0.05 or less. When a significant main effect was detected, post hoc Tukey tests were performed again at an alpha significance level of 0.05 on the least squares means subject by condition effect to determine where the condition main effect (normal, weighted or weight off) was.

Lastly, in order to answer the third research question discussed earlier, another two-way repeated measure ANOVA was performed on the normal obstacle condition trials from group 2 and the last 20 adapted obstacle trials from group 1 on the variable of toe obstacle clearance. Again here the two within factors of the ANOVA were condition and trial. However, in this case only two conditions existed of normal obstacle clearance and adapted obstacle clearance. Trials were still 1, 2, 3, 4, 5 and 6 because of the 20 trials in each condition, trials 6 through 20 were collapsed into one trial. Interaction and main effects were all looked for again in the same manner as stated above at an alpha significance level of 0.05. If effects existed, then post hoc Tukey tests were utilized in the same way as stated above as well.

Chapter 4

Experimental Results

4.0: Group 1 – Normal Walking Group

After manipulating the shank segment mass of the normal walking group, it was evident that changes to the walking patterns of the participants occurred. These changes were seen at the ankle, knee and hip joints and in both the conditions of weight and weight off. Some of these results were immediate but did not last for many trials, some lasted throughout all the experimental trials, some were only affected by the conditions of weight or weight off alone and in some cases no results were found. The results will be presented in the following order and only the variables which showed statistically significant effects will be presented:

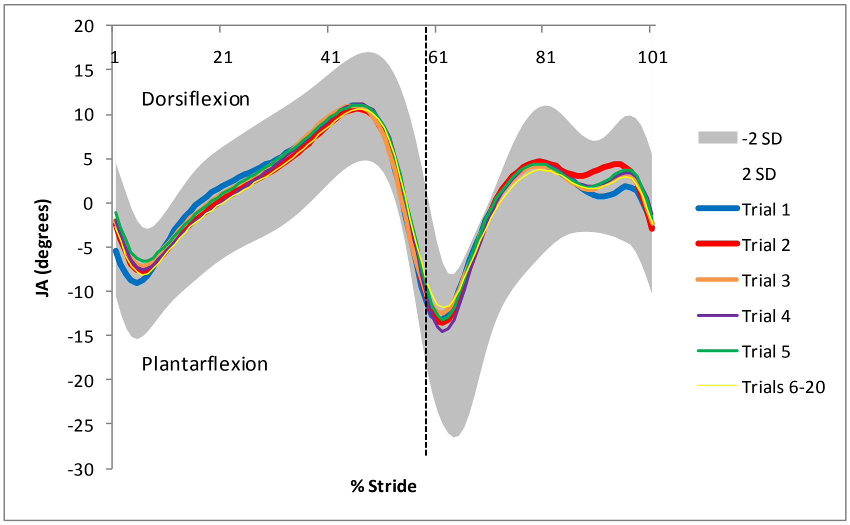
1. Time series data graphs of the weighted and weight off conditions at the ankle, knee and hip for joint angles, joint moments and joint powers through one full gait cycle.
2. Stance phase peaks in the order of joint angles, joint moments and joint powers.
3. Toe off specific results in the order of joint angles, joint moments, joint powers, toe velocity in the horizontal direction and toe velocity in the vertical direction.
4. Swing phase peaks in the order of joint angles, joint moments and joint powers.
5. Heel contact specific results in the order of joint angles, joint moments, joint powers, toe velocity in the horizontal direction and toe velocity in the vertical direction.
6. Minimum toe clearance through swing phase results.

4.1: Full Gait Cycle Graphs

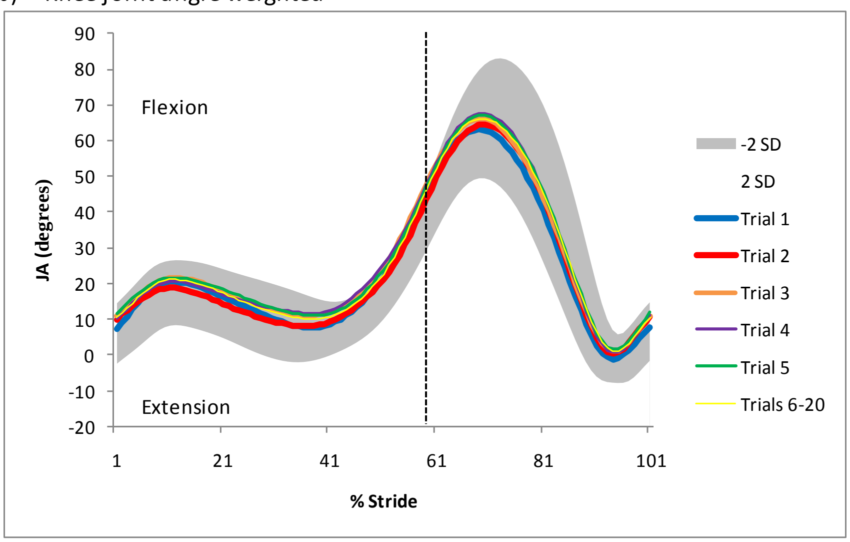
Below are the graphs (Figures 3 through 8) of the ankle, knee and hip joint angles, moments and powers during one full gait cycle of the weighted and weight off conditions compared to +/- 2 standard deviations (2SD) of the ten subjects in the normal walking condition. Each line represents the average of all ten subjects for a specific trial in each condition (trials 1 through 6). Again, trials 6 through 20 were binned together so these trials are all represented by a single line. The black line running through the graphs also shows where the stance phase ends and the swing phase starts. It is also very important to keep in mind the amount of variability that exists in normal human gait. Winter, DA (1991) states that ankle, knee and hip joint angles throughout the gait cycle are quite similar within one individual subject with less than 2 degrees of difference occurring between trials. Even between subjects, this variability is not very large. However, when looking at the ankle, knee and hip joint moments and powers throughout a gait cycle, the variability within and between subjects is much larger. This effect is more prevalent at the knee and hip. So, averaging the knee and hip joint moments and powers across trials and across subjects largely diminishes this variability seen.

In certain cases where the stance kinematics and kinetics dominated the scale of the full gait cycle graphs, the swing phase was graphed separately to the side from right toe off to right heel contact with an appropriate scale for clearer viewing.

a) Ankle joint angle weighted



b) Knee joint angle weighted



c) Hip joint angle weighted

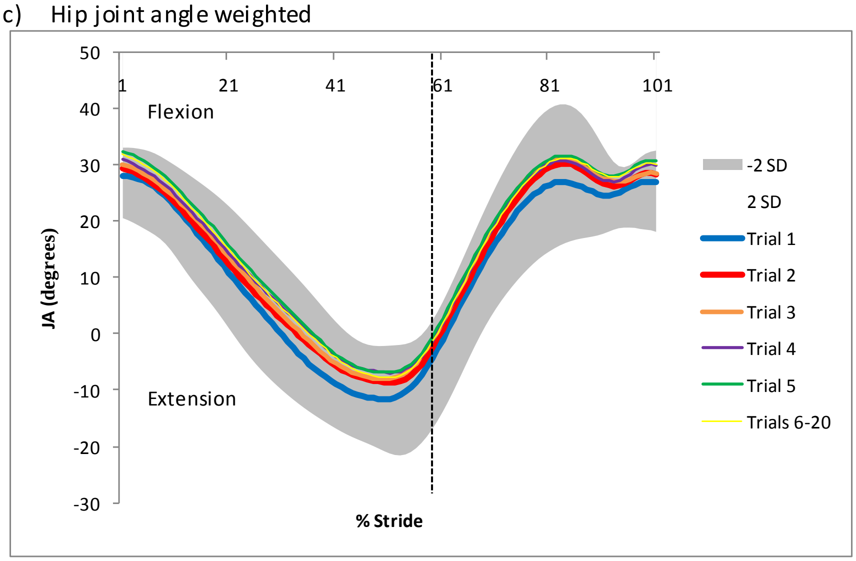
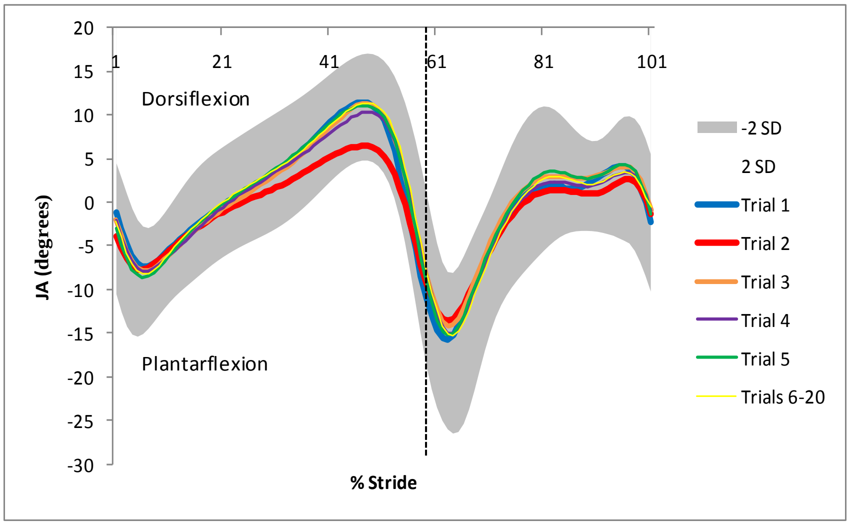
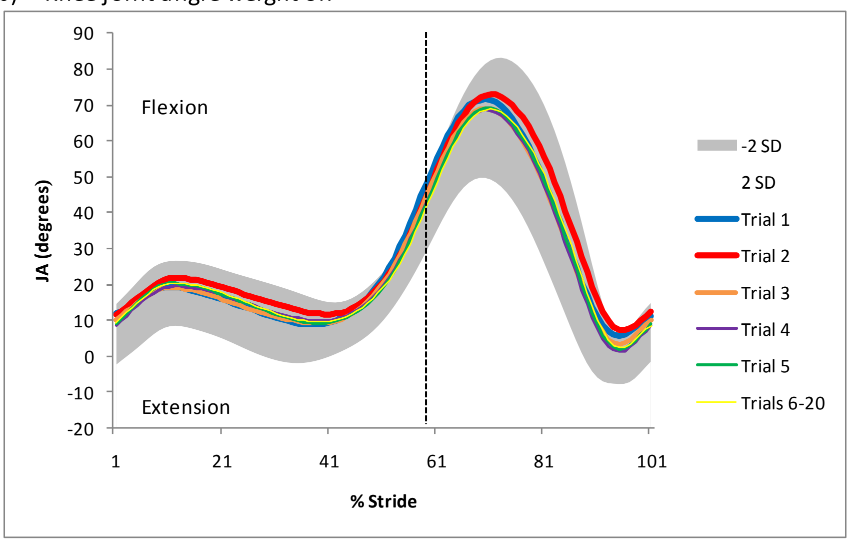


Figure 3: One full gait cycle of the weighted condition for the ankle (a), knee (b) and hip (c) joint angles.

a) Ankle joint angle weight off



b) Knee joint angle weight off



c) Hip joint angle weight off

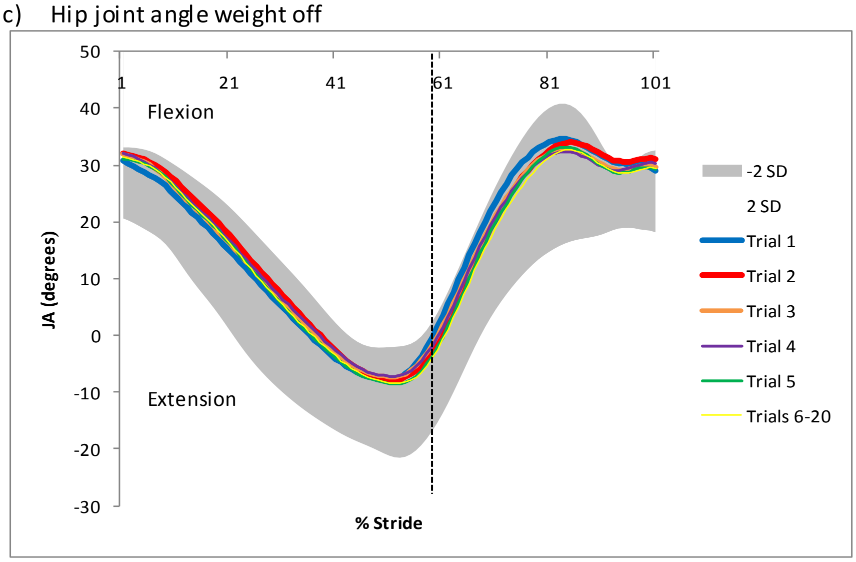


Figure 4: One full gait cycle of the weight off condition for the ankle (a), knee (b) and hip (c) joint angles.

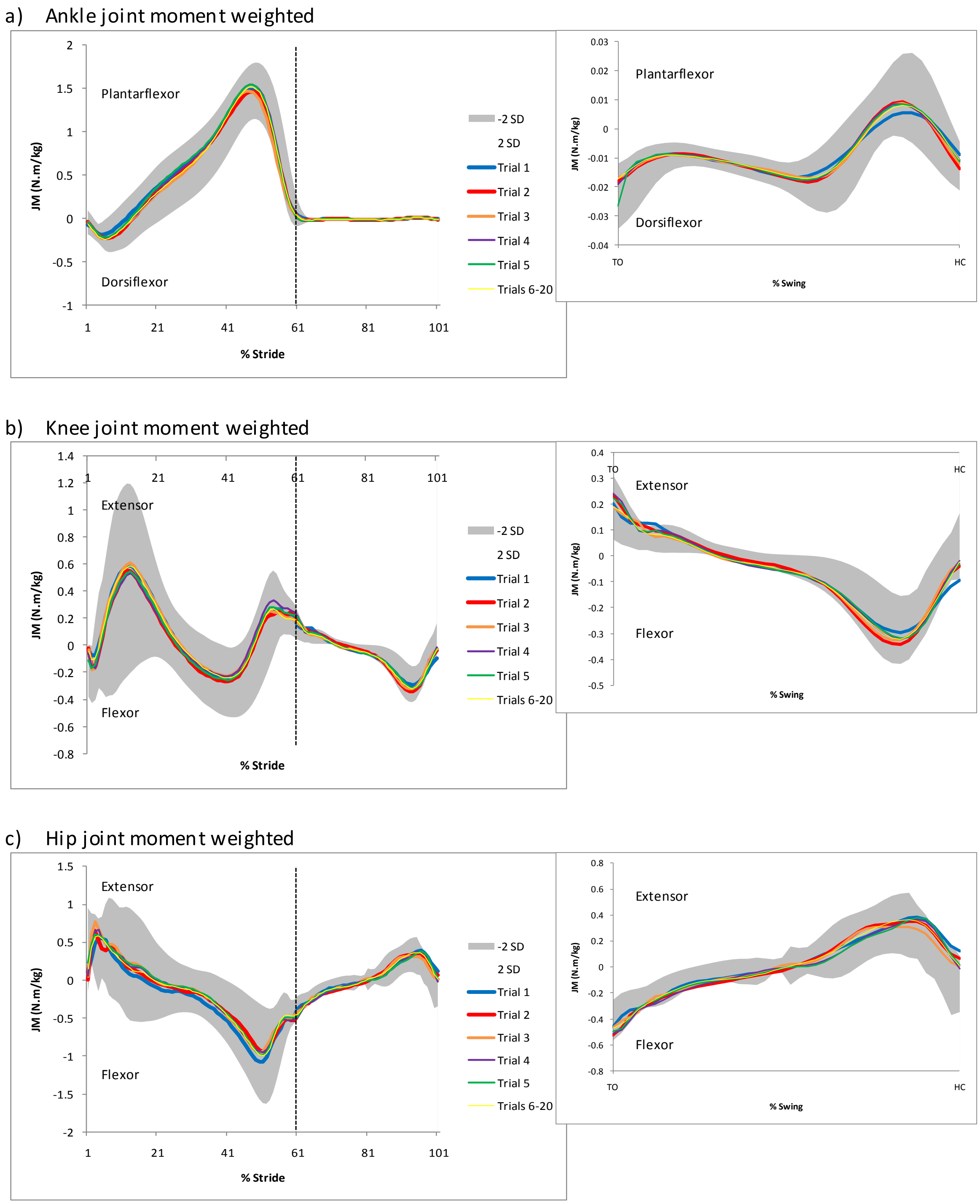
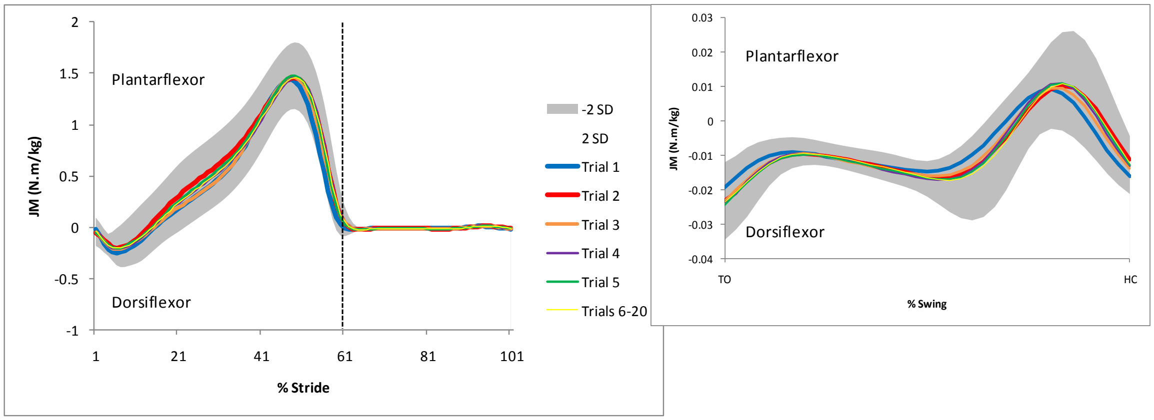
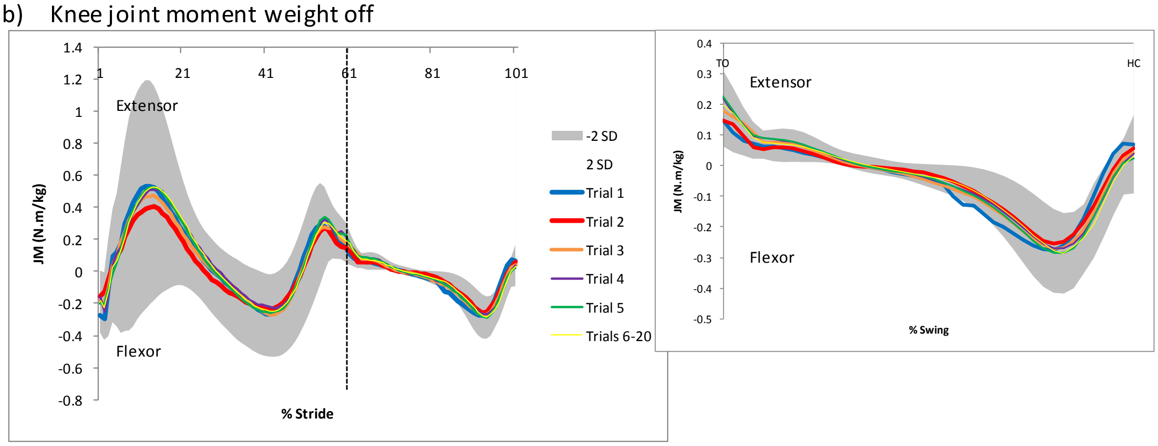


Figure 5: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint moments.

a) Ankle joint moment weight off



b) Knee joint moment weight off



c) Hip joint moment weight off

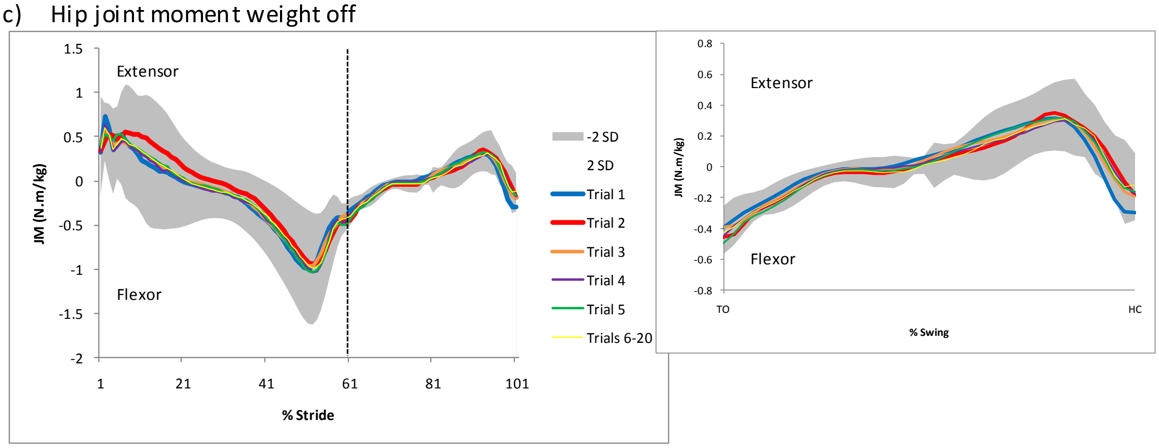
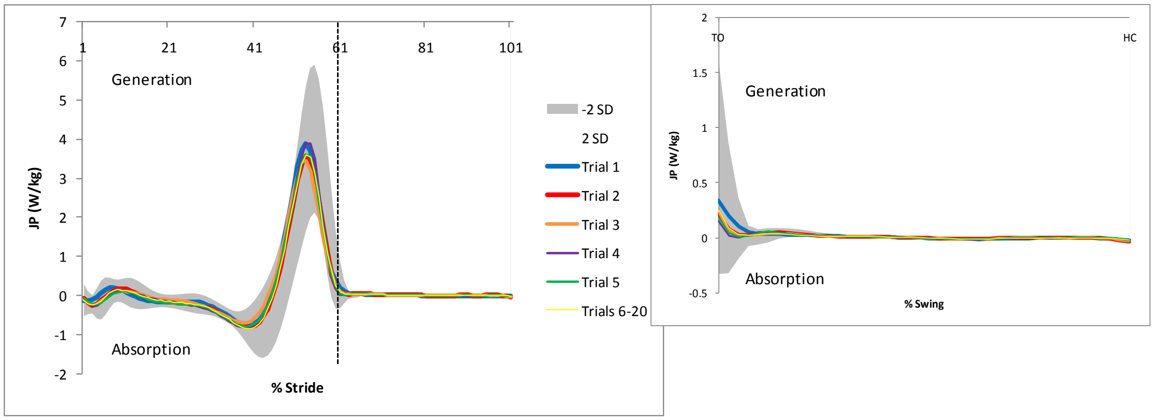
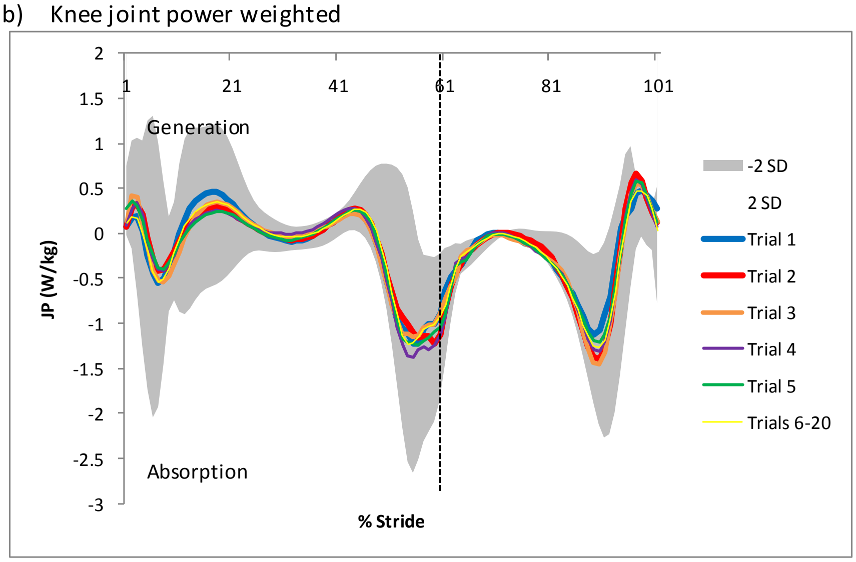


Figure 6: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint moments.

a) Ankle joint power weighted



b) Knee joint power weighted



c) Hip joint power weighted

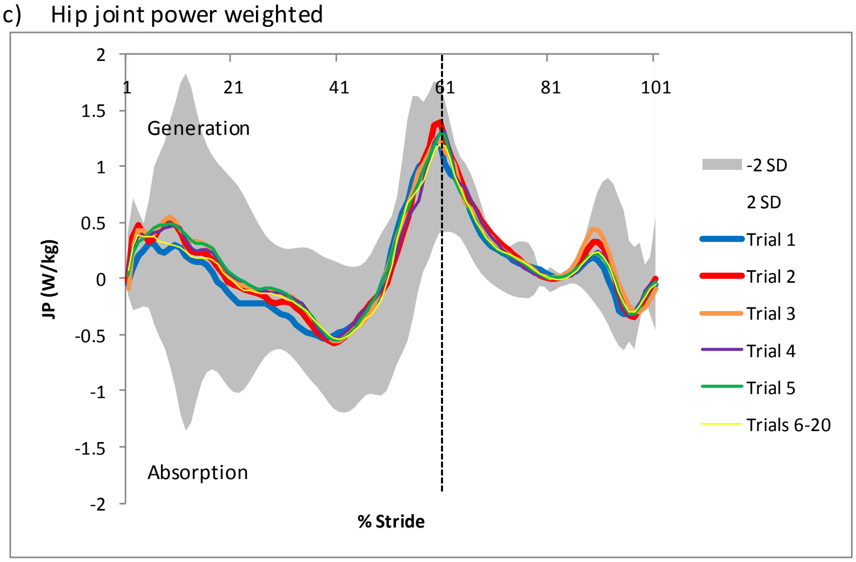
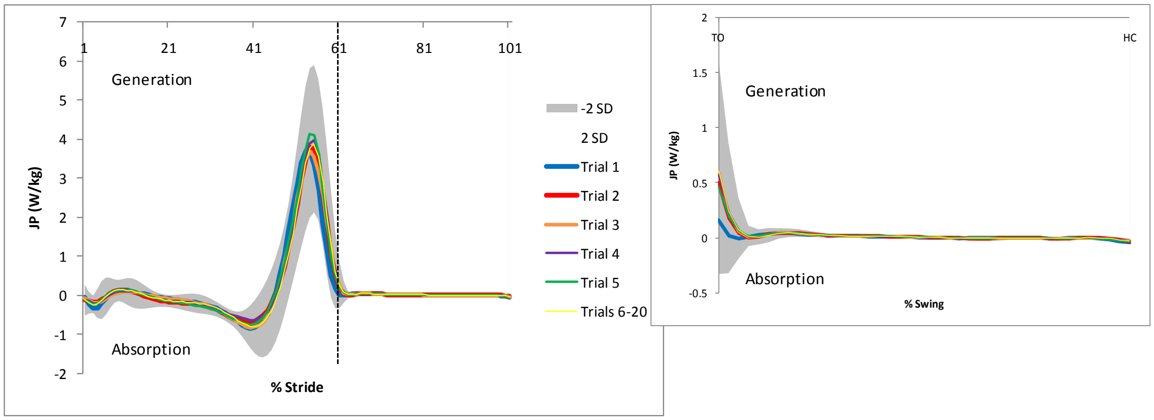
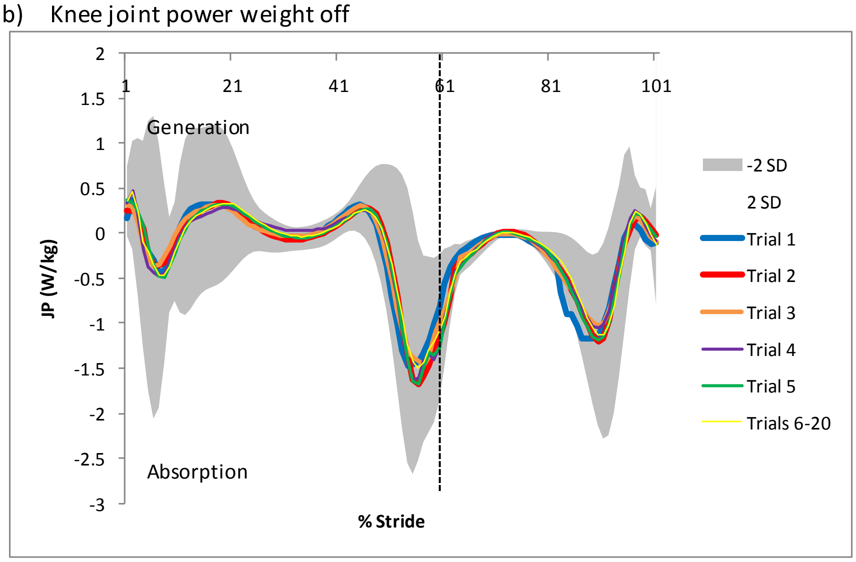


Figure 7: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint powers.

a) Ankle joint power weight off



b) Knee joint power weight off



c) Hip joint power weight off

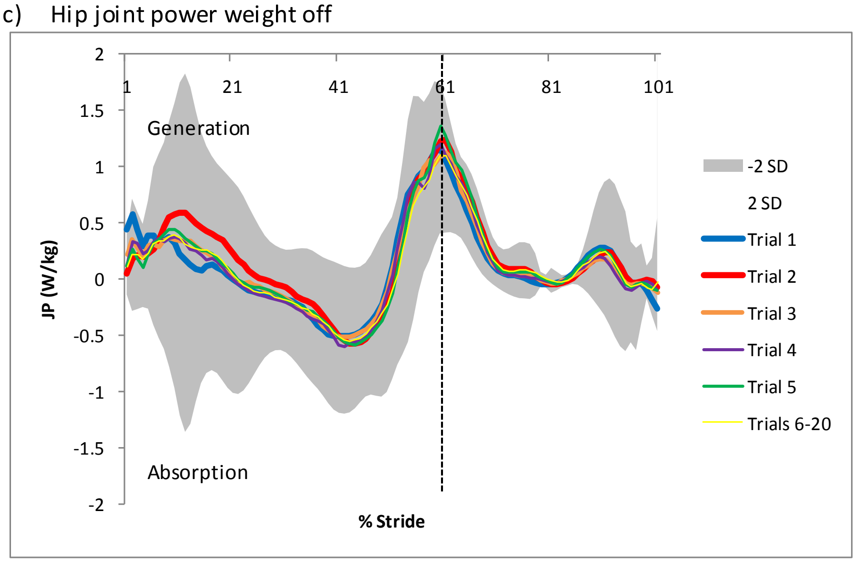


Figure 8: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint powers.

4.2: Stance Peak Results

At first sight there does not seem to be obvious differences between the normal walking, weighted walking and weight off conditions. However, the knee joint peak angle showed a condition main effect (F=11.00, p=0.0008). This knee joint angle main effect revealed that the weighted and weight off conditions produced increased knee flexion angles compared to normal walking. A trial by condition interaction effect however, was detected for the hip joint peak angle (F=2.29, p=0.0191). Post hoc Tukey tests determined two results. First, trial 1 of all three conditions produced hip joint extension angles different from each other with the weighted condition being decreased compared to normal and the weight off condition being even further decreased (Figure 9). Secondly, throughout trials 2 to 6 the weighted and weight off conditions produced decreased hip joint extension angles from normal but not different from each other (Figure 9).

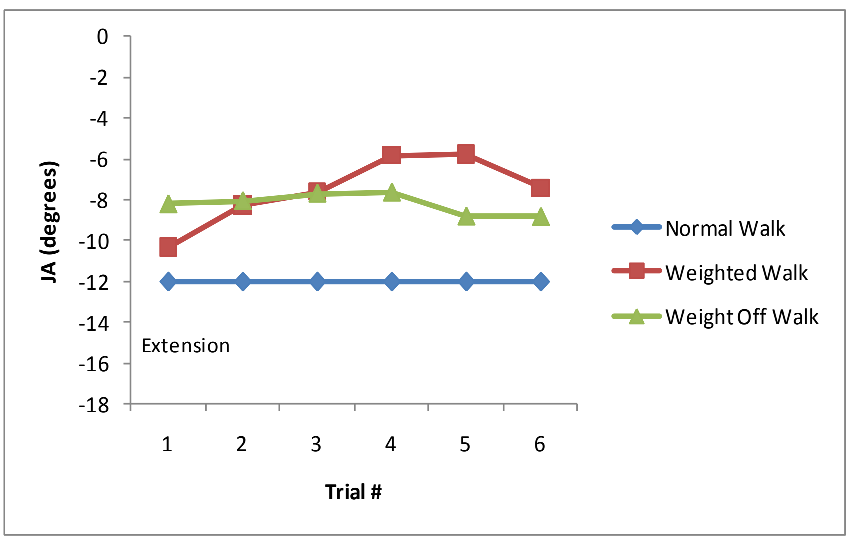


Figure 9: Hip joint peak angle during stance. This graph and the graphs like it to follow show the trial by condition interaction effects detected by statistical analysis. Each data point represents the average of 10 subjects for a particular discrete point analyzed in the gait cycle for a given variable (hip joint angle during stance in this case) with each particular experimental condition (normal, weighted and weight off) and trial (1 through 6) being shown separately. Trial 6 represents trials 6 through 20 binned together and then averaged.

Both the knee (F=3.85, p=0.0407) and hip (F=21.74, p=<0.0001) joint moment showed condition main effects. The joint extensor moment at the knee was increased in the weight off condition compared to normal and increased even more so in the weighted condition. The same was true for the hip flexor moments.

During stance, the hip joint power variable was the only one to show a trial by condition interaction effect (F=3.32, p=0.001). Again two results were detected by post hoc Tukey tests. First, in trial 1 all three conditions produced hip joint powers different from each other with the weighted condition generating more energy compared to normal and the weight off condition energy generation being even further increased (Figure 10). Secondly, throughout trials 2 to 6 the weighted and weight off conditions had increased hip joint energy generation from normal but not different from each other (Figure 10).

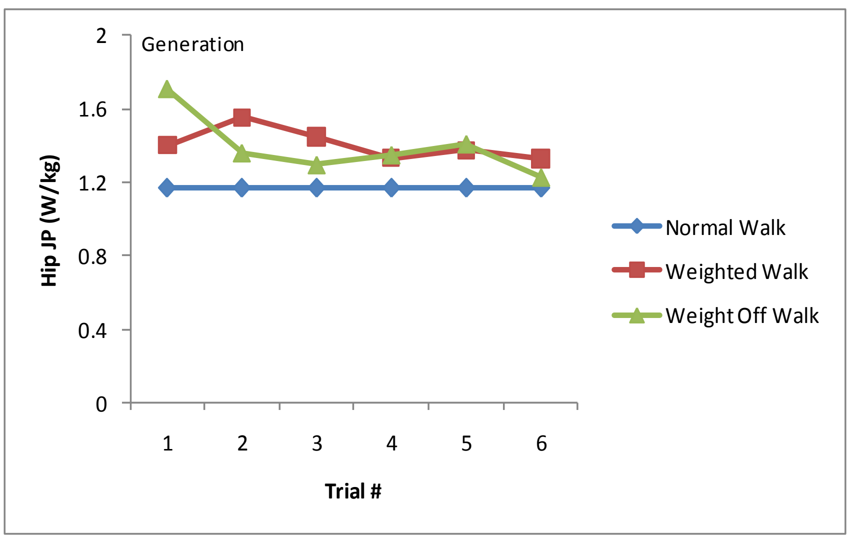


Figure 10: Hip joint peak power during stance

Shown in Appendix D (i) is a summary of all the statistical results seen during stance phase of the gait cycle for the ankle, knee and hip joint angles, moments and powers.

4.3: Toe Off Results

When the right toe off data points in the gait cycle are looked at more closely, there are some significant differences. Starting with the joint angles, the repeated measures ANOVA revealed a condition by trial interaction significant effect for the ankle joint angle variable with an F-value of 2.16 and a p-value of 0.0272. After performing the post hoc Tukey tests on this interaction effect, it was clear that the ankle plantarflexion angles at toe off were decreased in the weight off condition compared to the normal condition for trials 1, 2, 3 and 6 (Figure 11). Additionally, the ankle plantarflexion angles were decreased in the weighted condition compared to the normal condition for trials 3 and 6 (Figure 11).

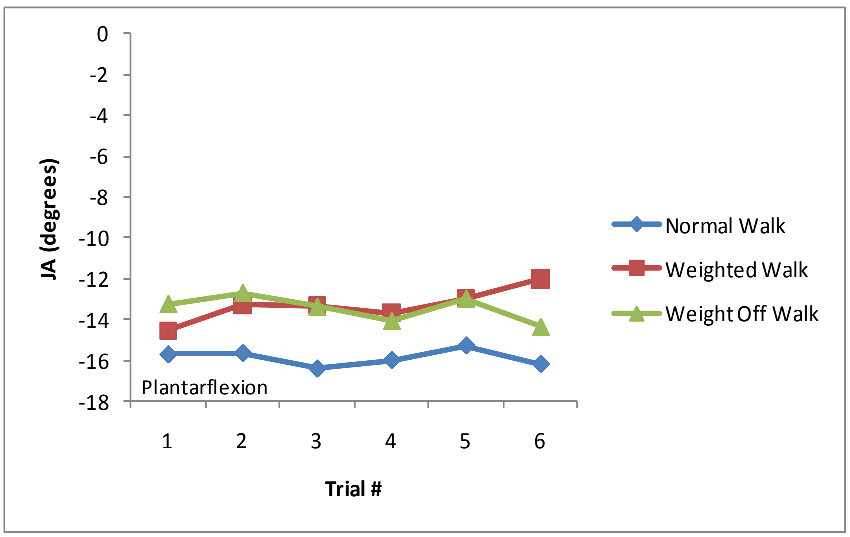


Figure 11: Ankle joint angles at toe off

For the hip joint angle, a significant trial by condition interaction effect (F=4.15, p=<0.0001) was found. Post hoc Tukey tests performed on the hip joint angle at toe off revealed that the weighted and weight off conditions were different from normal. The weighted condition had decreased hip extension angles for trials 1 through 5 and even started to flex in trial 6 (Figure 12). In the weight off condition, the hip was flexed in trials 1 through 5 (Figure 12).

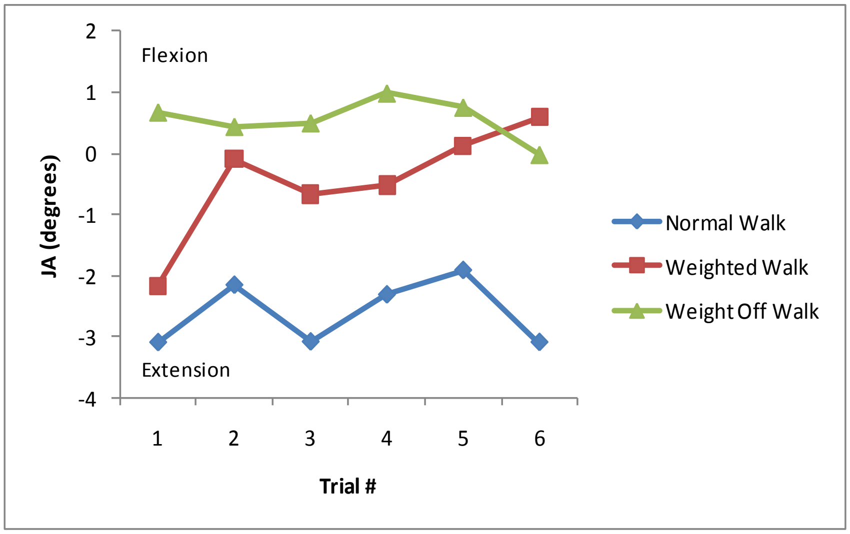


Figure 12: Hip joint angles at toe off

Moving on to the joint moments, like the ankle joint angle at toe off, a significant trial by condition interaction was found for the ankle joint moment at toe off (F=2.27, p=0.0203) as well. Post hoc Tukey tests revealed that the weighted walking condition produced a greater ankle joint plantarflexor moment than the normal and weight off conditions. This was seen in the significant differences between the normal and weighted conditions in trials 1, 3, 4 and 6 (Figure 13) and trials 1, 2, 3 and 6 (Figure 13) between the weighted and weight off conditions.

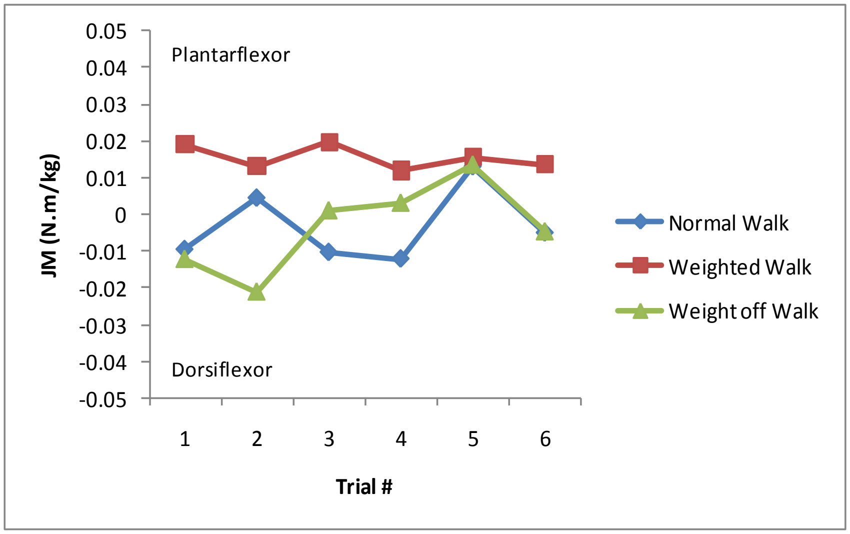


Figure 13: Ankle joint moments at toe off

A condition main effect was found for the knee joint moment (F=11.46, p=0.0006) where the weighted condition produced increased knee extensor moment compared to the normal and weight off conditions. The result of the hip joint moment at toe off showed a similar result to that of the knee. Again, a significant condition main effect was found (F=13.66, p=0.0002) showing that the hip joint flexor moment was increased in the weighted condition compared to the normal and weight off condition.

Finally, the ankle joint power results at toe off showed the same trend as the ankle angle and ankle moment at toe off. The repeated measures ANOVA detected a significant trial by condition effect for ankle joint power (F=2.03, p=0.0393). Post hoc Tukey tests did not identify a very clear pattern of exactly where the specific changes occurred between trials and conditions. However, most notably a significant difference was seen between trial 1 normal walking and trial 1 weighted walking (Figure 14) where the weighted ankle joint power generation was decreased and actually absorbed energy.

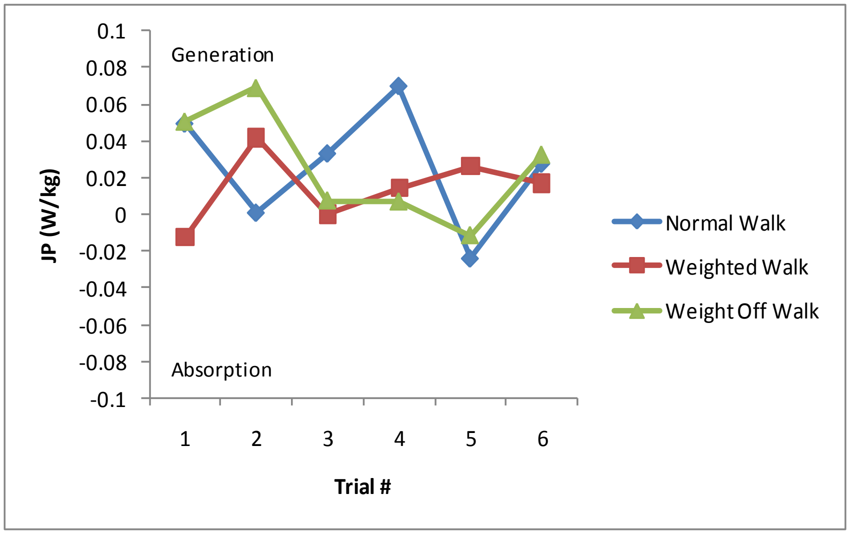


Figure 14: Ankle joint powers at toe off

At toe off, significant condition main effects were found at the knee and hip for joint powers. The condition main effect for knee joint power (F=6.8, p=0.0063) showed an increase in energy absorption and the hip joint power (F=8.74, p=0.0022) showed an increase in energy production throughout the weighted and weight off conditions compared to that of the normal walking condition.

No significant interaction or main effects were seen for the toe velocity in the horizontal direction (progression) of the right leg at toe off. However, toe velocity in the vertical direction (longitudinal axis) of the right leg was found to have a significant trial by condition interaction effect (F=2.08, p=0.0338). Post hoc Tukey tests showed that there was always a significant difference between weighted walking versus normal walking and weight off walking. Specifically, the toe velocity during weighted walking was always decreased compared to that of the other conditions across all trials (Figure 15).

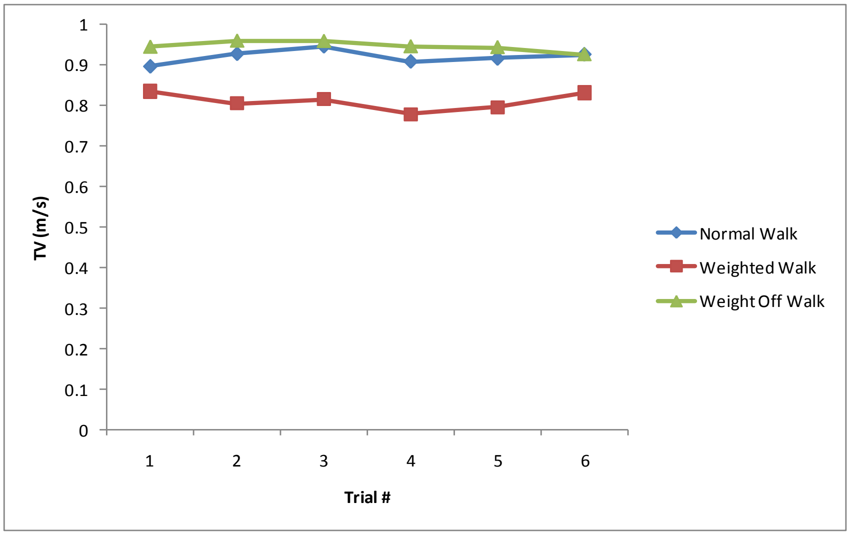


Figure 15: Toe velocity in the vertical direction at toe off

In order to visualize these normal walking group results at toe off more easily and view all the results together as a group see Appendix D (ii). It shows all the variables discussed above at toe off in a summarized fashion.

4.4: Swing Peak Results

The peak ankle angle during swing showed a condition main effect ($F=14.43$, $p=0.0002$) where all the conditions were different from each other. The weight off condition showed increased ankle dorsiflexion angles compared to normal and the weighted condition ankle dorsiflexion angles were even more increased.

At the knee, a trial by condition interaction effect was detected for the knee joint peak angle during swing ($F=5.12$, $p=<0.0001$). Post hoc Tukey tests revealed two separate results. Initially, in the first 2 trials, all three conditions produced knee joint angles different from each other. The weighted condition knee flexion joint angles were decreased compared to normal while the weight off condition knee flexion joint angles were increased compared to normal (Figure 16). In the remaining 4 trials (3 to 6) the weighted condition knee angles returned to values similar to in the normal condition but the weight off condition knee angles remained increased compared to normal (Figure 16).

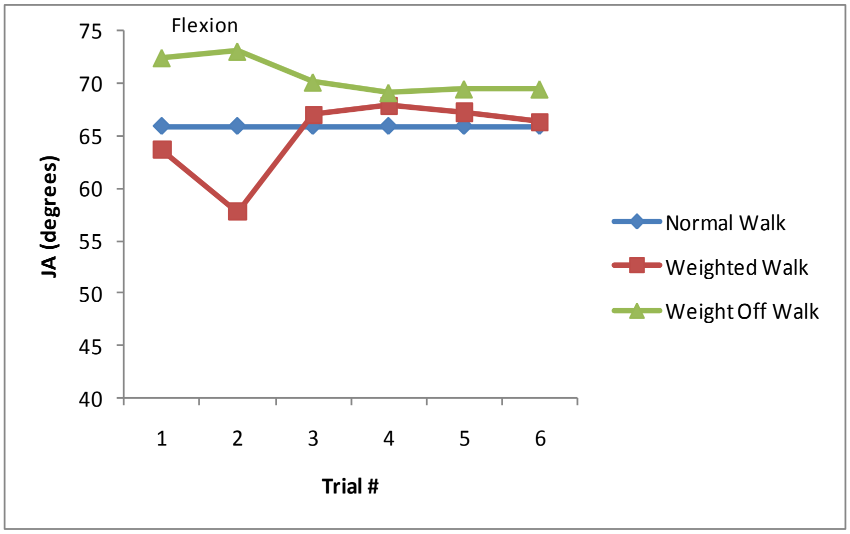


Figure 16: Knee joint peak angle during swing

Like the knee, a significant trial by condition interaction effect was also found for the hip joint peak angle during swing ($F=3.56$, $p=0.0005$). Post hoc Tukey tests revealed significant differences between all three conditions across all six trials (Figure 17). Both the weighted and weight off conditions produced increased hip joint flexion angles with the biggest increases being in the weight off condition (Figure 17).

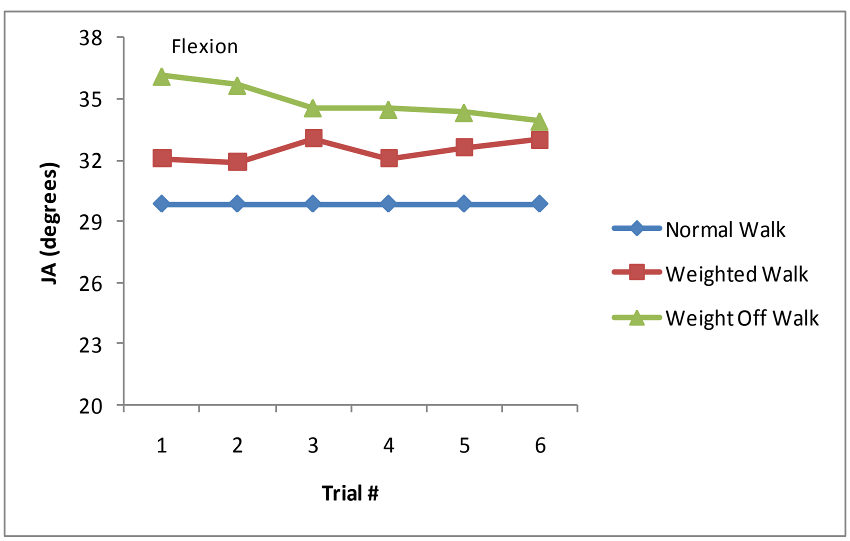


Figure 17: Hip joint peak angle during swing

Looking at the joint moment peaks during the swing phase of the gait cycle, condition main effects were detected for the ankle, knee and hip joints. The condition main effect for the peak ankle joint moment ($F=3.81$, $p=0.0418$) showed that the weighted condition produced an increased ankle joint plantarflexor moment compared to normal. The condition main effects for the knee and hip were ($F=8.83$, $p=0.0021$) and ($F=7.59$, $p=0.0041$) respectively. It was determined that the knee joint flexor moment was increased while the hip joint extensor moment was increased in the weighted condition compared to the normal and weight off conditions.

Similar to the knee joint above, the joint peak powers during the swing phase of the gait cycle were also all found to show condition main effects at the ankle, knee and hip joints. The ankle joint peak power condition main effect ($F=5.63$, $p=0.0126$) showed that the weighted condition had decreased joint energy generation compared to normal. The knee joint peak power condition main effect ($F=4.55$, $p=0.0251$) showed an increase in energy absorption in the weighted condition. At the hip, the joint peak power condition main effect ($F=11.82$, $p=0.0005$) showed an increase in joint energy generation for both the weighted and weight off conditions but with no difference between each other. So at the ankle and knee, a decrease in joint power was detected, but at the hip an increase in joint power was found instead.

Appendix D (iii) summarizes the statistical results found for the joint angles, moments and powers at the ankle, knee and hip joints for their peak values during the swing phase of the gait cycle.

4.5: Heel Contact Results

At the knee, only a significant condition main effect was found. The condition main effect had an F-value of 8.63 and a p-value of 0.0024. Post hoc Tukey tests showed that the knee flexion joint angles in the weighted and weight off conditions were increased compared to the normal walking condition. However, the hip joint showed a significant trial by condition interaction effect at heel contact (F=3.19, p=0.0015). Post hoc Tukey tests revealed that the weighted condition had increased hip flexion angles in trials 2 through 6 (Figure 18) and the weight off condition was the same but for trials 1 through 6 (Figure 18).

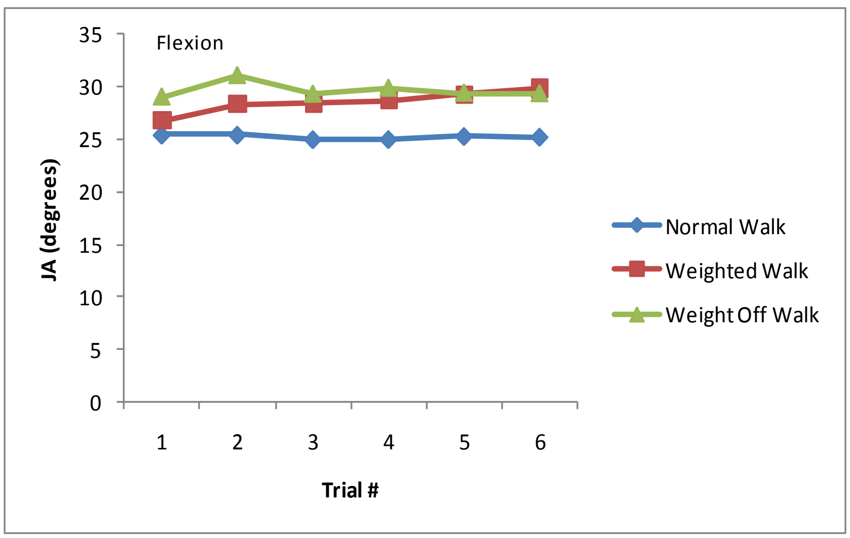


Figure 18: Hip joint angles at heel contact

The knee moment at heel contact however, did show a trial by condition interaction effect (F=2.34, p=0.0165). The post hoc Tukey tests revealed that these differences between conditions and trials occurred between all trials across all conditions (Figure 20). These differences showed that the weighted walking condition had a decreased knee joint flexor moment compared to the normal walking condition and the weight off condition actually had a knee extensor joint moment (Figure 19).

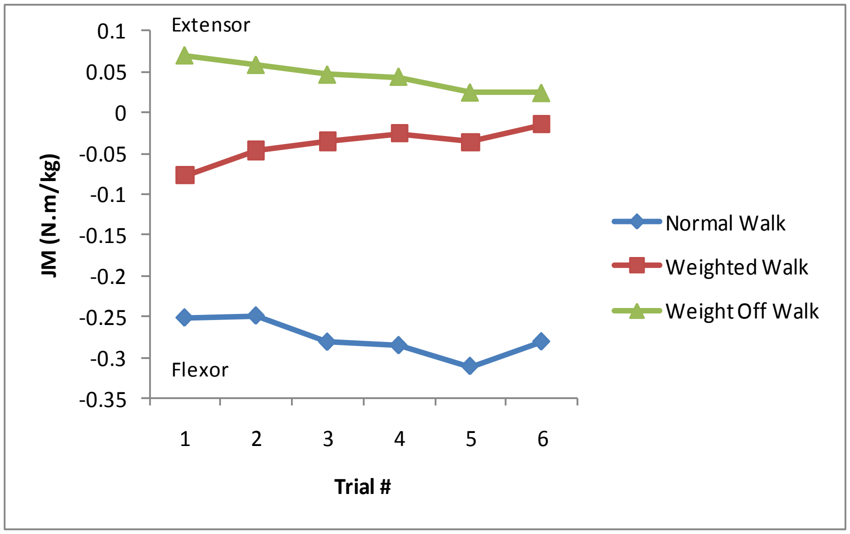


Figure 19: Knee joint moments at heel contact

Similarly, the repeated measures ANOVAs test found a significant trial by condition interaction effect (F=3.78, p=0.0003) for the hip joint moment at heel contact. The post hoc Tukey tests also revealed that these differences between trials and conditions occurred in all trials across all conditions (Figure 20). At the hip however, the joint flexor moment was decreased in the weight off condition compared to normal and in the weighted condition the hip moment became an extensor moment (Figure 20).

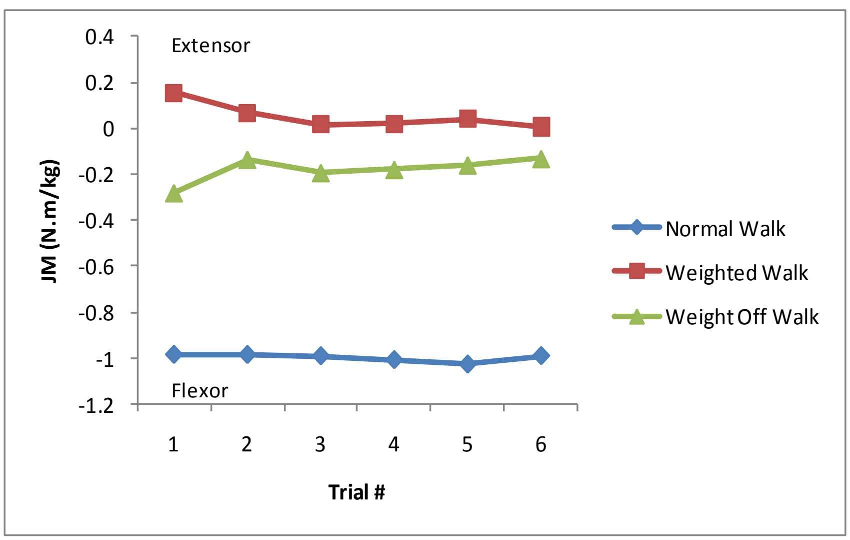


Figure 20: Hip joint moments at heel contact

Continuing on to the joint powers at heel contact, the only statistical results found was at the knee. The knee joint power showed a significant trial by condition interaction effect (F=1.98, p=0.0447). Post hoc Tukey tests found that the only significant difference seen between trials across conditions occurred in trial 1 between the weighted and weight off condition. Figure 21 shows this result and also how the weighted knee joint energy generation at heel contact spikes in trial 1 before leveling back down to similar absorption values in the normal and weight off conditions over the remaining five trials.

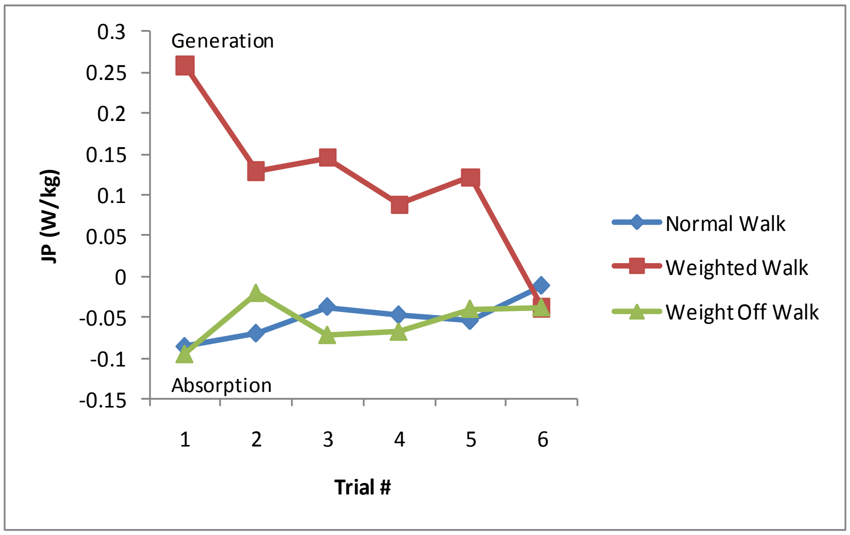


Figure 21: Knee joint powers at heel contact

Finally, the toe velocities of the right great toe at heel contact revealed no significant trial by condition interaction effects from the repeated measures ANOVAs for toe velocity in the horizontal or vertical direction. However, significant condition main effects were found in both the horizontal direction (F=11.46, p=0.0006) and vertical direction (F=3.61, p=0.0479). Specifically, it was found that in the weighted condition, the toe velocities at heel contact were increased in both directions.

Appendix D (iv) displays all the results discussed above for all the variables analyzed at heel contact.

4.6: Minimum Toe Clearance

The last variable analyzed in the normal walking group was minimum toe clearance and the repeated measures ANOVA test detected a trial by condition interaction effect (F=3.76, p=0.0003). The post hoc Tukey tests showed two results. First, in trial 1 only, the weight off condition produced an increased minimum toe clearance compared to all other trials across conditions (Figure 22). Secondly, in trials 3 through 6, the weighted condition produced increased minimum toe clearances when compared to all other trials across conditions (Figure 22).

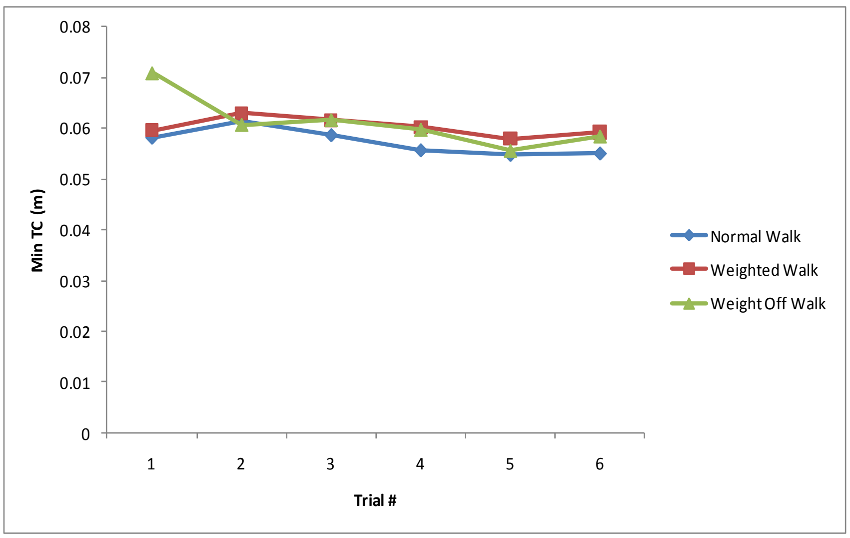


Figure 22: Minimum Toe Clearance

4.7: Group 2 – Obstacle Clearance Group

After manipulating the shank segment mass of the obstacle clearance group, it was evident that changes to the obstacle clearance patterns of the participants had also occurred. These changes were seen at the ankle, knee and hip joints and in both the conditions of weight and weight off. Some of these results were immediate but did not last for many trials, some lasted throughout all the experimental trials, some were only affected by the conditions of weight or weight off alone and in some cases no results were found. The results will be presented in the same order as was the normal walking group and only the variables which showed statistically significant effects will be presented:

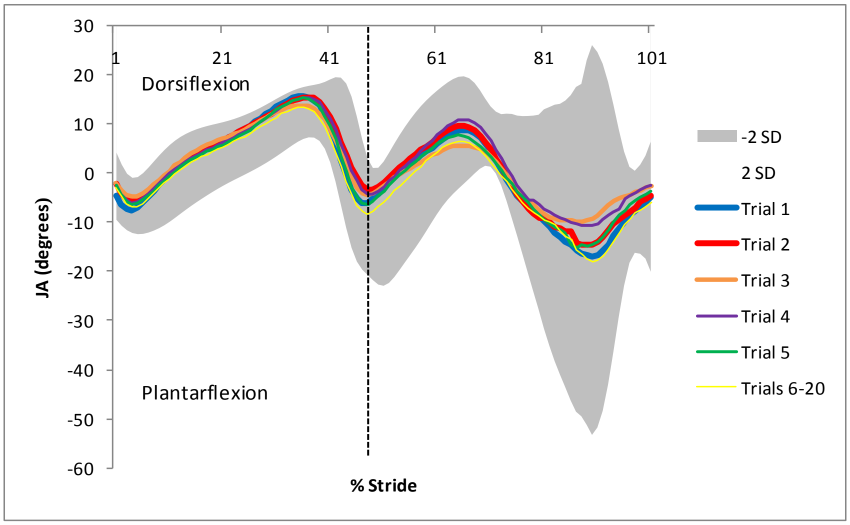
1. Time series data graphs of the weighted and weight off conditions at the ankle, knee and hip for joint angles, joint moments and joint powers through one full gait cycle.
2. Stance phase peaks in the order of joint angles, joint moments and joint powers.
3. Toe off specific results in the order of joint angles, joint moments, joint powers, toe velocity in the horizontal direction and toe velocity in the vertical direction.
4. Swing phase peaks in the order of joint angles, joint moments and joint powers.
5. Heel contact specific results in the order of joint angles, joint moments, joint powers, toe velocity in the horizontal direction and toe velocity in the vertical direction.
6. Toe obstacle clearance results.

4.8: Full Gait Cycle Graphs

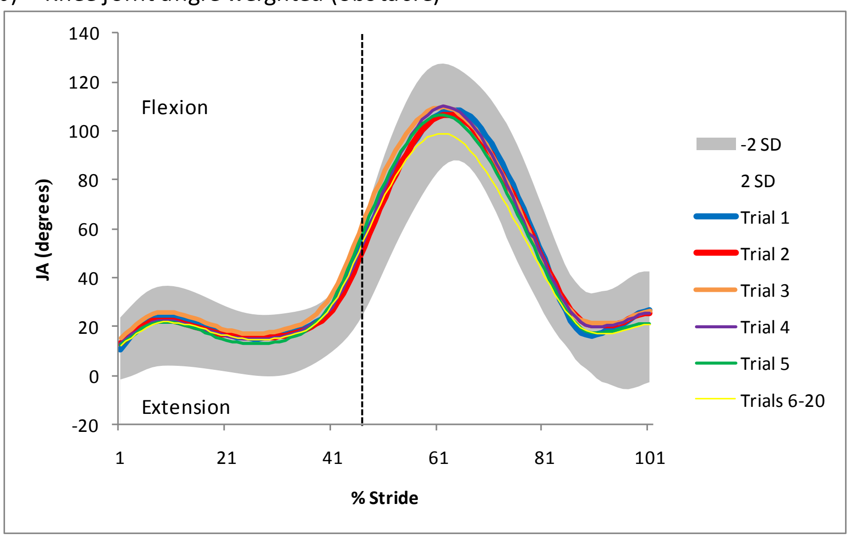
Below are the graphs (Figures 23 through 28) of the ankle, knee and hip joint angles, moments and powers during one full gait cycle of the weighted and weight off conditions compared to +/- 2 standard deviations (2SD) of the ten subjects in the normal obstacle clearance condition. Each line represents the average of all ten subjects for a specific trial in each condition (trials 1 through 6). Again, trials 6 through 20 were binned together so these trials are all represented by a single line. The black line running through the graphs also shows where the stance phase ends and the swing phase starts.

In certain cases where the stance kinematics and kinetics dominated the scale of the full gait cycle graphs, the swing phase was graphed separately to the side from right toe off to right heel contact with an appropriate scale for clearer viewing.

a) Ankle joint angle weighted (obstacle)



b) Knee joint angle weighted (obstacle)



c) Hip joint angle weighted (obstacle)

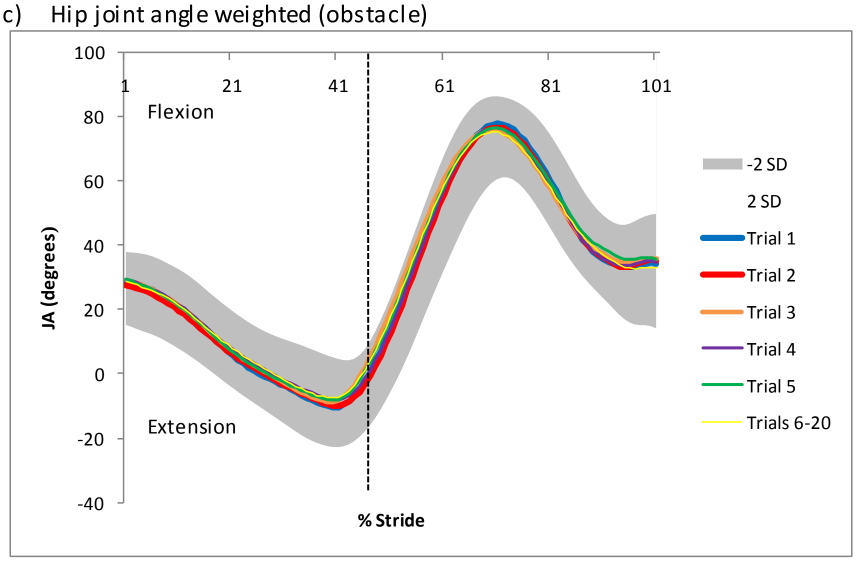
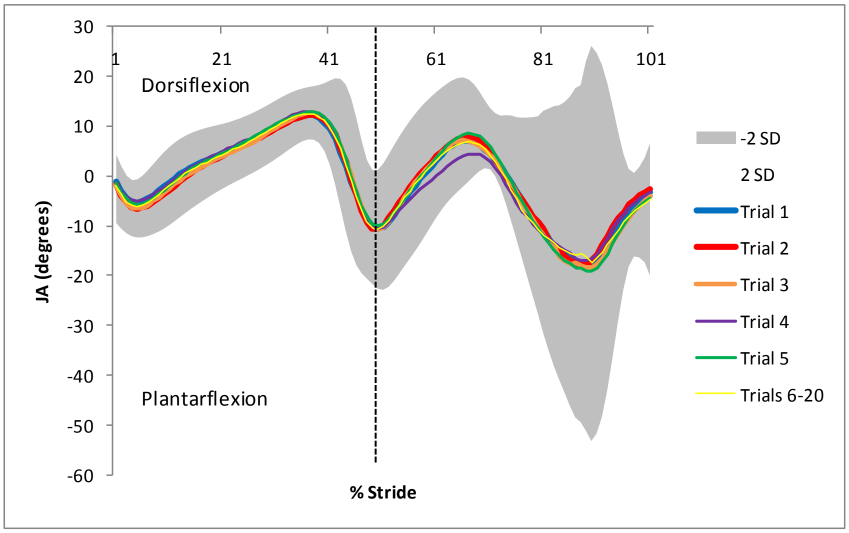
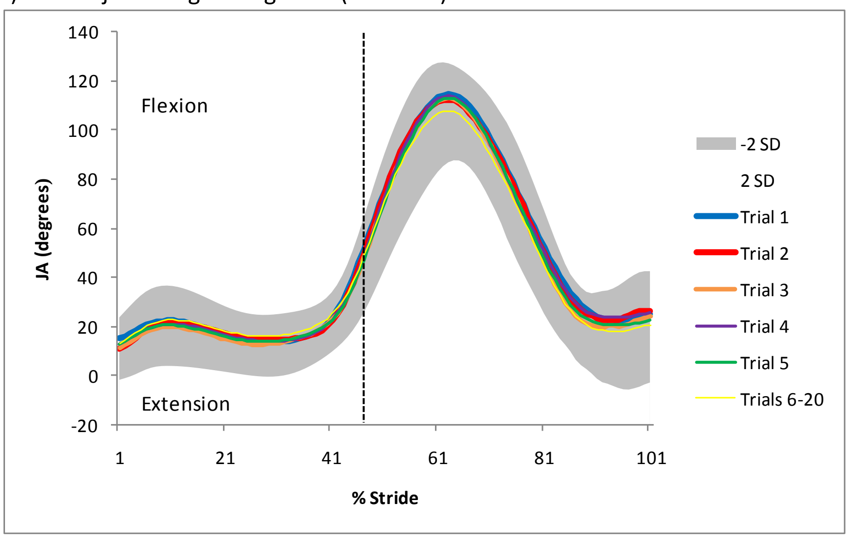


Figure 23: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint angles.

a) Ankle joint angle weight off (obstacle)



b) Knee joint angle weight off (obstacle)



c) Hip joint angle weight off (obstacle)

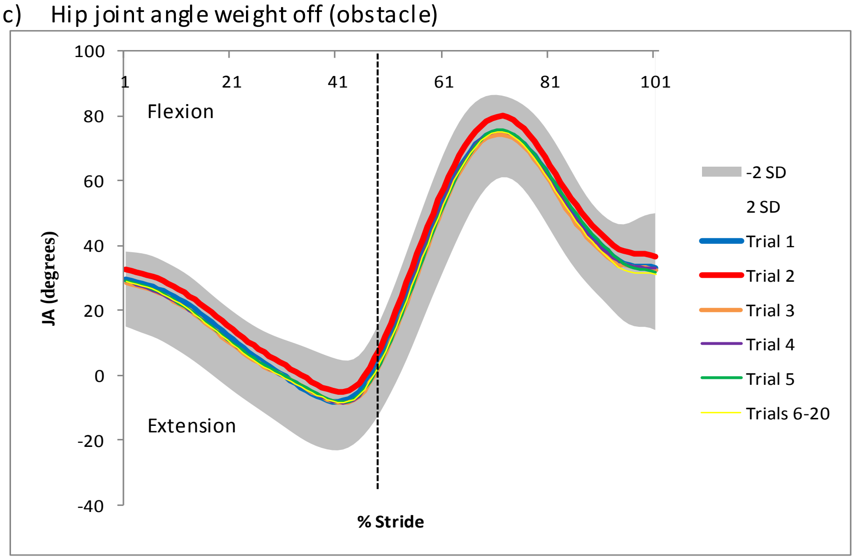
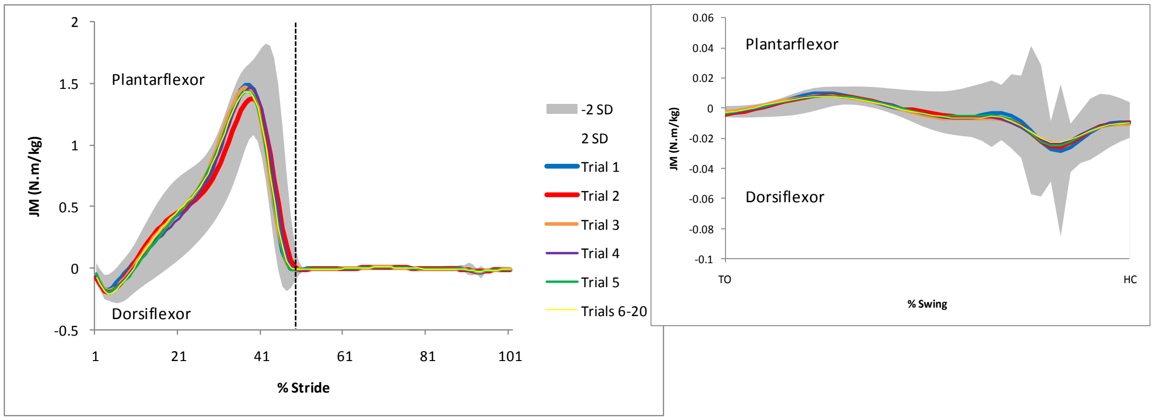
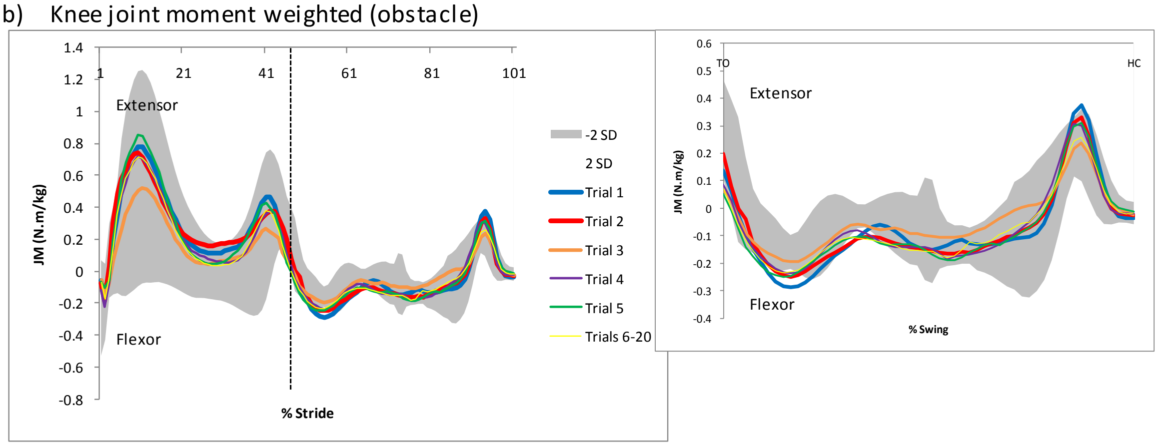


Figure 24: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint angles.

a) Ankle joint moment weighted (obstacle)



b) Knee joint moment weighted (obstacle)



c) Hip joint moment weighted (obstacle)

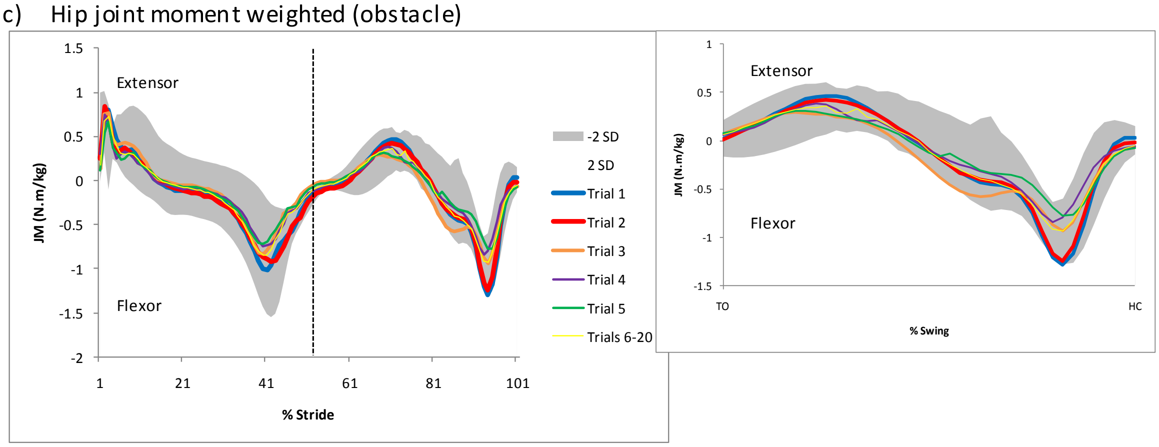


Figure 25: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint moments.

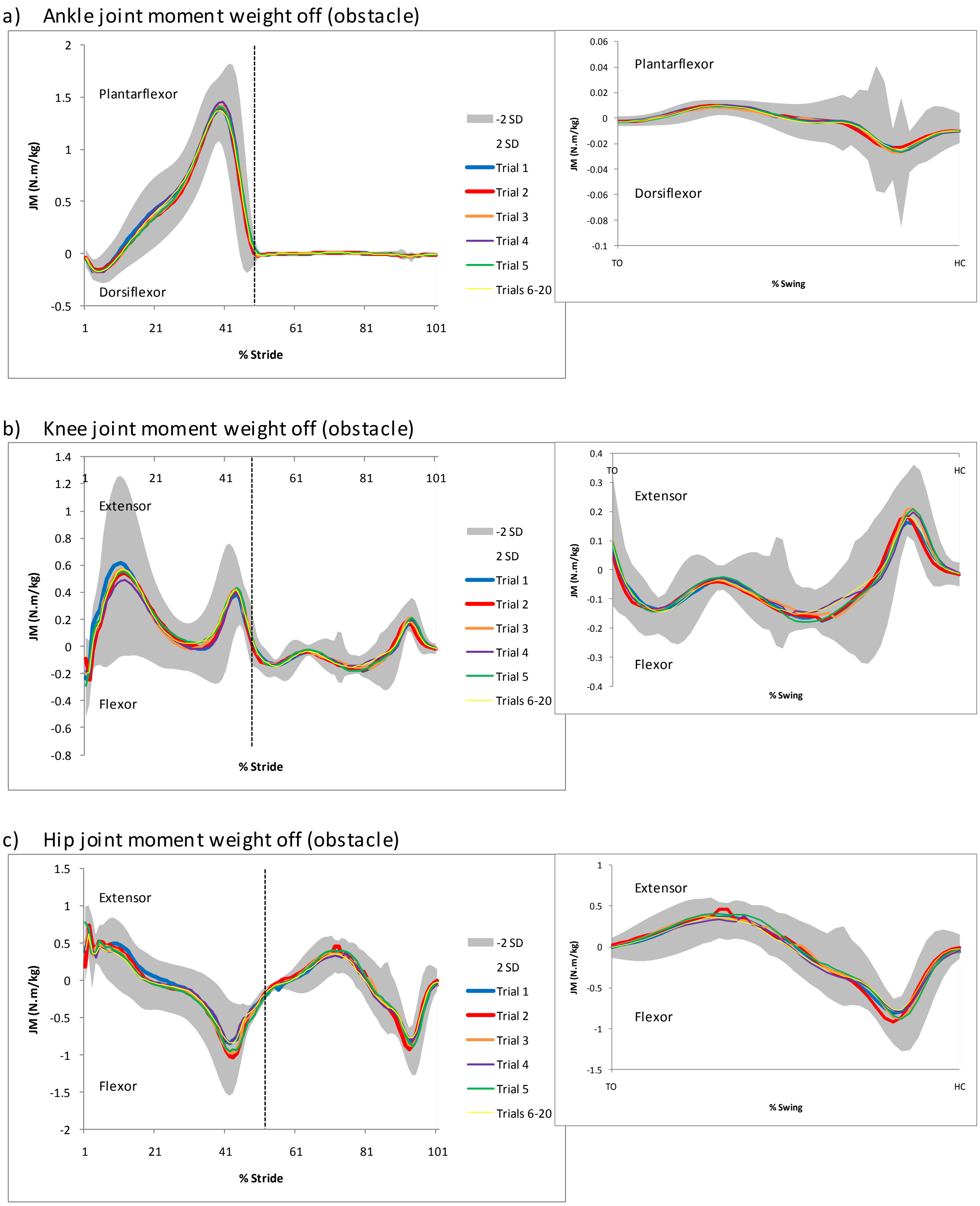
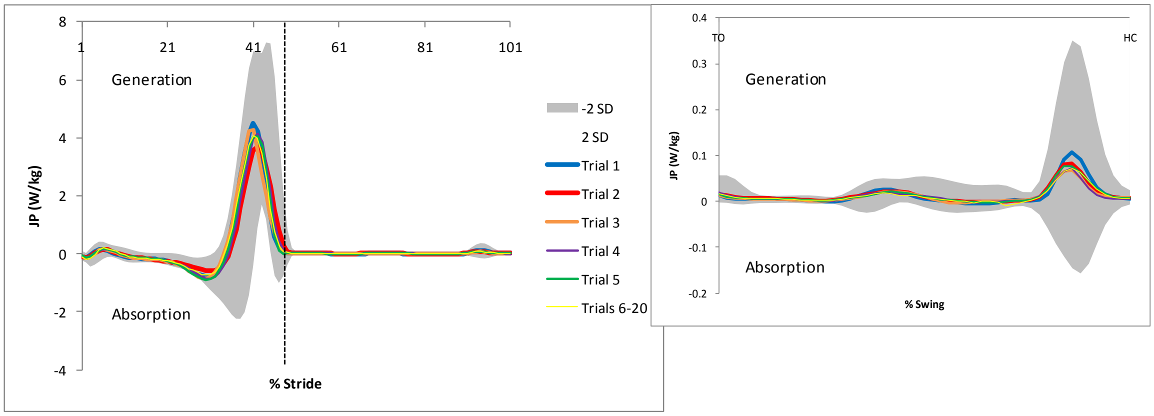
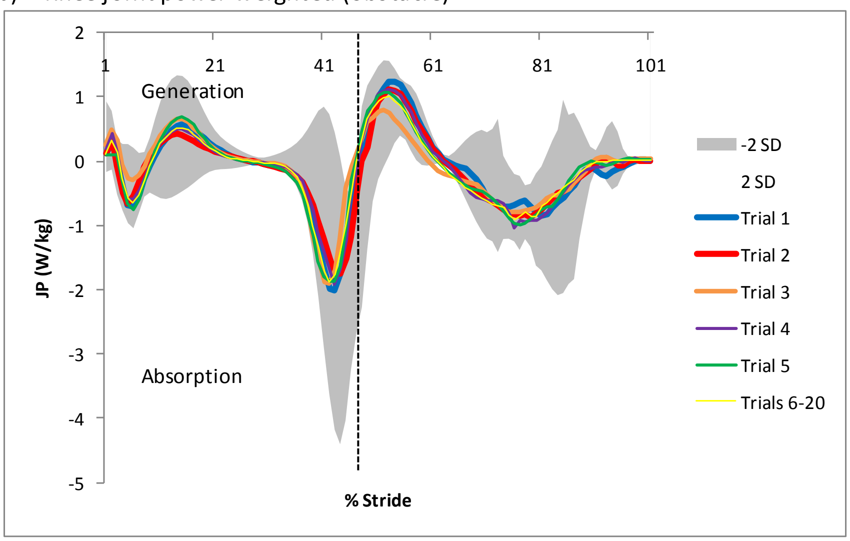


Figure 26: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint moments.

a) Ankle joint power weighted (obstacle)



b) Knee joint power weighted (obstacle)



c) Hip joint power weighted (obstacle)

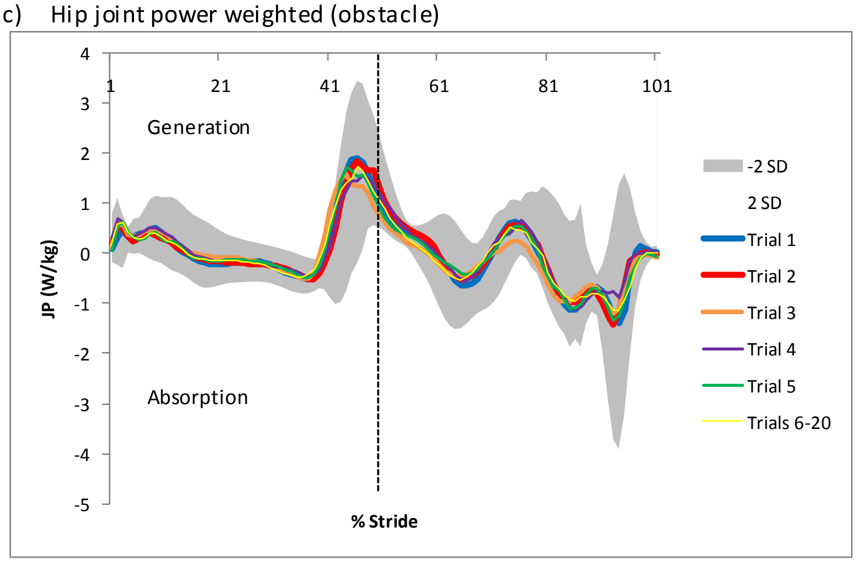
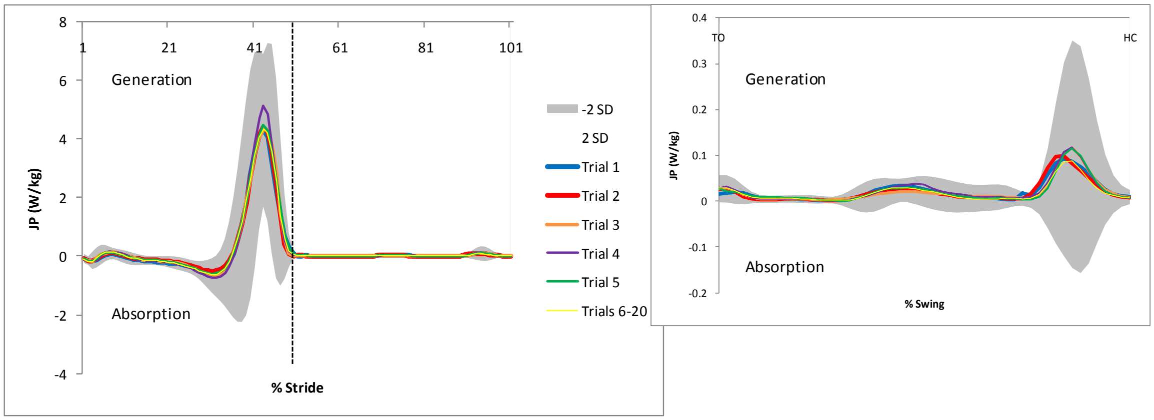
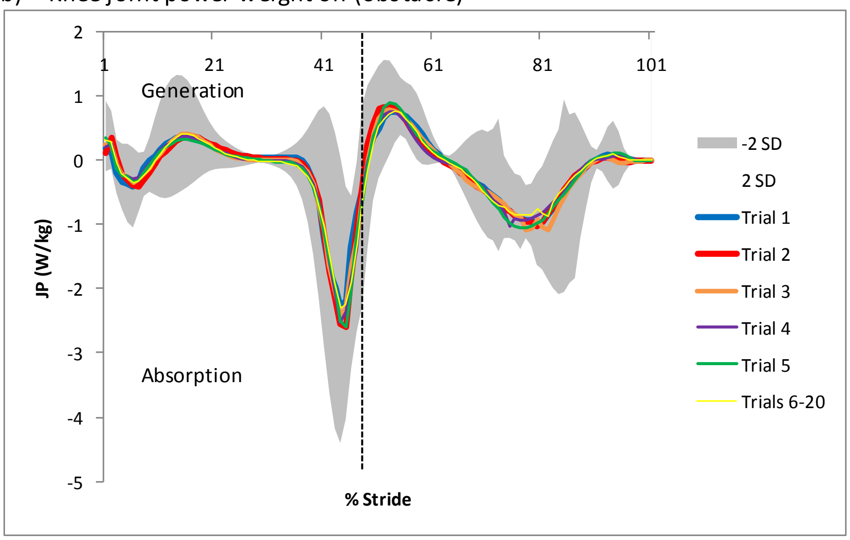


Figure 27: One full gait cycle of the weighted condition for ankle (a), knee (b) and hip (c) joint powers.

a) Ankle joint power weight off (obstacle)



b) Knee joint power weight off (obstacle)



c) Hip joint power weight off (obstacle)

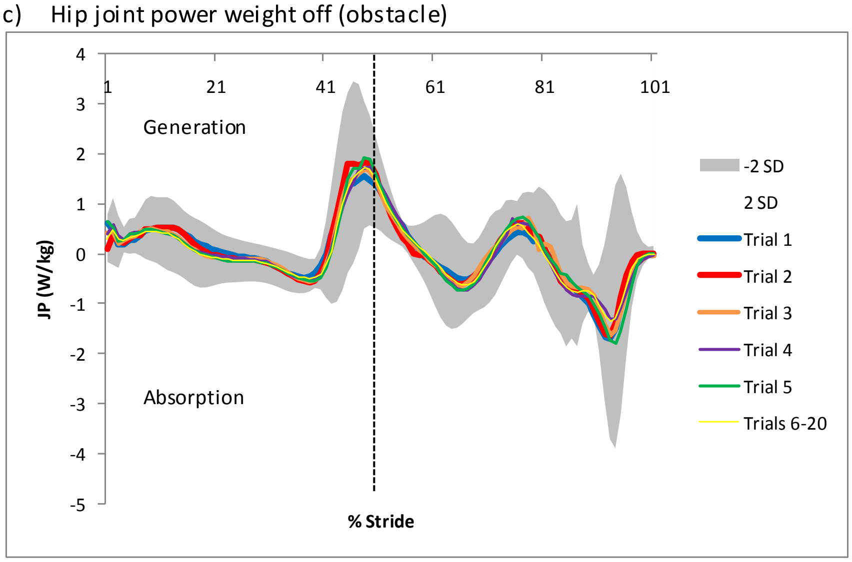


Figure 28: One full gait cycle of the weight off condition for ankle (a), knee (b) and hip (c) joint powers.

4.9: Stance Peak Results

In the stance phase of the obstacle clearance group, a trial by condition interaction effect was detected for the ankle joint peak angle ($F=5.75$, $p=<0.0001$). The post hoc Tukey tests revealed that the weighted obstacle clearance condition produced increased ankle joint dorsiflexion angles compared to the normal and weight off conditions throughout all trials (Figure 29).

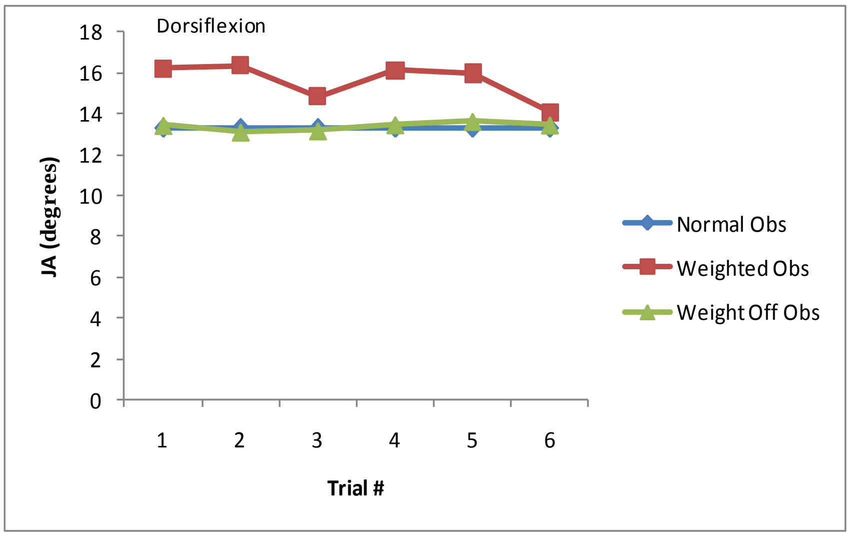


Figure 29: Ankle joint peak angle during stance (obstacle). This graph and the graphs like it to follow show the trial by condition interaction effects detected by statistical analysis. Each data point represents the average of 10 subjects for a particular discrete point analyzed in the gait cycle for a given variable (ankle joint angle during stance in this case) with each particular experimental condition (normal, weighted and weight off) and trial (1 through 6) shown separately. Trial 6 represents trials 6 through 20 binned and averaged together.

The knee joint peak angle during the stance phase of the obstacle clearance group was also determined to show a significant condition main effect ($F=5.88$, $p=0.0108$). This result revealed that the knee joint flexion angles in the weighted obstacle clearance condition were increased compared to the normal condition.

When looking at the joint moments during the stance phase of the obstacle clearance group, condition main effects were found at the ankle, knee and hip. The ankle joint moment condition main effect ($F=15.79$, $p=0.0001$) showed that the weighted obstacle clearance condition produced larger ankle joint plantarflexor moments compared to normal. The knee joint moment condition main effect (F=11.02, p=0.0008) showed that the weighted condition had an increased knee joint extensor moment compared to the normal and weight off conditions. Finally, the hip joint moment condition main effect (F=6.71, p=0.0067) showed that the weighted and weight off obstacle clearance conditions had increased flexor moments compared to normal. In any case, the added weight seems to have a major effect on the joint moment at all three joints in the lower limb.

Just like the joint moments, the joint powers during the stance phase of the obstacle clearance group showed condition main effects at the ankle, knee and hip as well. The ankle joint power condition main effect (F=5.84, p=0.0376) determined that the weight off condition had increased ankle joint power generation compared to normal. At the knee, the condition main effect (F=18.25, p=<0.0001) revealed that the knee joint increased energy absorption in the weight off condition compared to the normal and weighted conditions. Lastly, the hip joint power condition main effect (F=6.53, p=0.0074) showed that the weighted and weight off condition had increased hip joint power generation compared to normal. Again, it seems that the added weight has a large effect on the joint powers at all three joints in the lower limb.

Appendix D (v) summarizes all these statistical results discussed above obtained in the obstacle clearance group during the stance phase of the gait cycle.

4.10: Toe Off Results

At toe off, no trial by condition interaction effects were found for the ankle joint angle variable but a significant condition main effect (F=6.19, p=0.009) was detected.

This condition main effect revealed that the ankle joint plantarflexion angles in the weighted obstacle clearance condition were significantly decreased compared to those in the normal or weight off obstacle clearance conditions. A significant condition main effect was also found for the hip joint angles at toe off (F=7.26, p=0.0049). This result showed that in the weight off obstacle clearance condition, the hip joint had increased flexion angles while in the normal and weighted conditions extension angles were present.

Of the joint moment variables analyzed at toe off, only a significant trial by condition interaction effect (F=2.65, p=0.0071) for the knee joint moment was found. Post hoc Tukey tests determined that the knee joint extensor moment in trial 1 of the weight off obstacle clearance condition was significantly decreased compared to those in the normal and weighted conditions (Figure 30). Oddly, this result disappeared in trials 2 and 3 but then existed again in trial 4 (Figure 30).

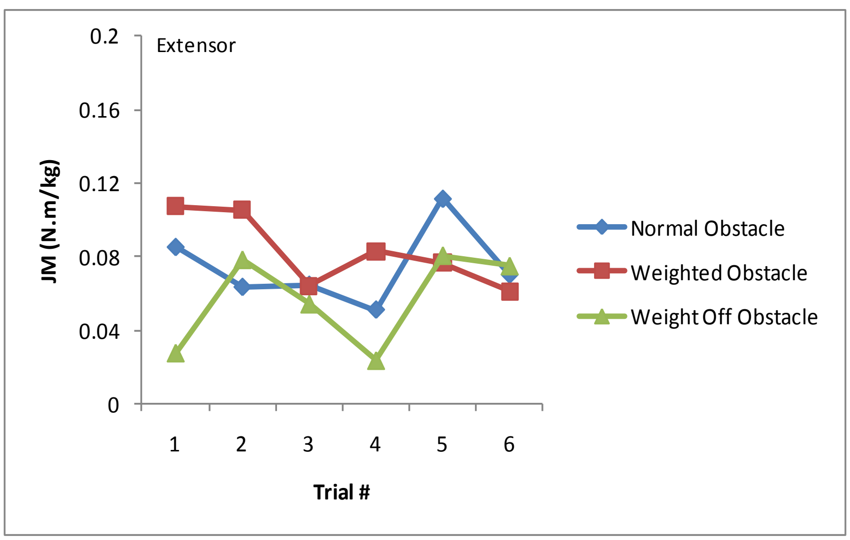


Figure 30: Knee joint moments at toe off (obstacle)

Like the joint moments, statistical analysis detected a knee joint power trial by condition interaction effect (F=2.71, p=0.0059) at toe off as well. After performing the post hoc Tukey tests, it was determined that in trial 1 of the weight off obstacle clearance condition, energy absorption decreased compared to the normal and weighted conditions. Graphically, the knee joint power in trial 1 of the weight off condition was increased compared to the other conditions (Figure 31).

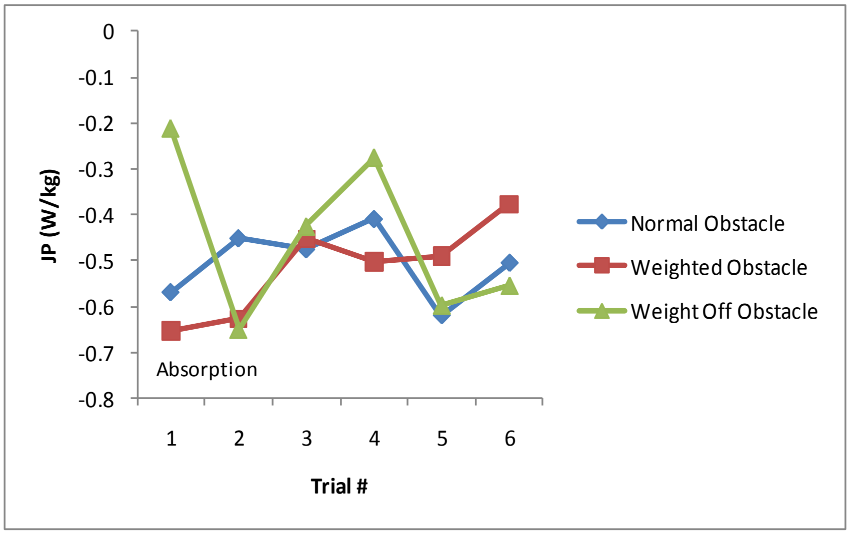


Figure 31: Knee joint powers at toe off (obstacle)

Similar to the toe velocity results in group 1 normal walking, the right toe velocity in the horizontal direction (progression) at toe off was not found to have any significant interaction or main effects results. However, also like in the normal walking group, the right toe velocity in the vertical direction (longitudinal axis) at toe off was determined to have a significant trial by condition interaction effect ($F=7.72$, $p=<0.0001$). After performing post hoc Tukey tests, it was determined that, in trial 1, there were significant differences between all three conditions (normal, weighted and weight off). This result showed that the weighted condition toe velocity was decreased compared to normal while the weight off toe velocity was increased compared to normal. Then in trials 2 through 5, only the weighted condition was significantly different from the other two conditions (Figure 32). Specifically, it should be noted that the weighted condition produced decreased toe velocities in the vertical direction until trial 6 where all three conditions became similar once again (Figure 32).

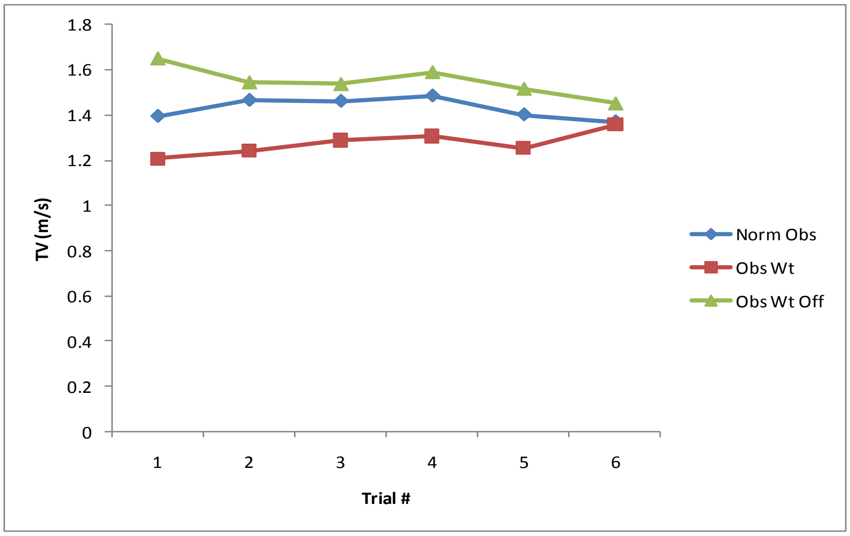


Figure 32: Toe velocity in the vertical direction at toe off (obstacle)

Shown in Appendix D (vi) is a summary sheet of all the statistical results found in the obstacle clearance group at toe off for joint angles, moments and powers at the ankle, knee and hip joints.

4.11: Swing Peak Results

During the swing phase of the obstacle clearance group all of the joints showed some significant statistical results. The ankle joint peak angle showed a condition main effect (F=7.34, p=0.029) where the weighted obstacle clearance condition had decreased ankle joint plantarflexion angles compared to the normal and weight off conditions. The knee joint peak angle variable was found to have a significant trial by condition interaction effect (F=2.58, p=0.0085). Post hoc Tukey tests revealed that the weight off obstacle clearance condition produced increased knee joint flexion angles compared to normal in trials 1 through 6 and compared to the weighted condition in trials 1, 2, 5 and 6 (Figure 33). Lastly, the hip joint peak angle also showed a condition main effect

(F=13.06, p=0.0003) where all three experimental conditions were different from each other. The weight off obstacle clearance condition had increased hip joint flexion angles compared to normal and the weighted condition flexion angles were even further increased from the weight off condition.

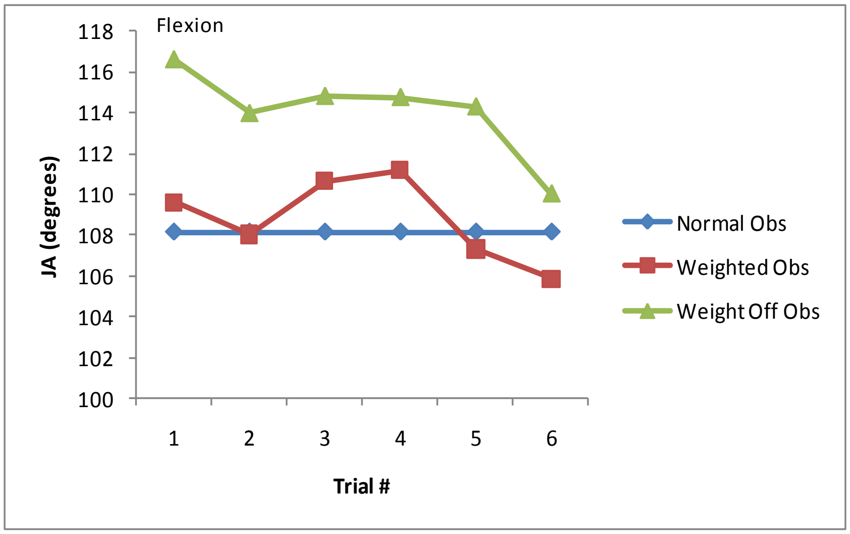


Figure 33: Knee joint peak angle during swing (obstacle)

Looking at the joint moments, both the knee (F=18.7, p=<0.0001) and hip (F=11.46, p=0.0006) joint moments were found to have condition main effects. The peak joint extensor moment of the knee and the peak joint flexor moment of the hip were increased in the weighted obstacle clearance condition compared to the normal and weight off conditions.

A summary of all the statistical results found at the ankle, knee and hip during the swing phase of the obstacle clearance group can be found in Appendix D (vii).

4.12: Heel Contact Results

Starting with joint angles, a significant trial by condition interaction effect was detected for the ankle joint angle at heel contact (F=1.97, p=0.0464). Post hoc Tukey tests revealed that in trials 1, 2 and 3 the ankle joint plantarflexion angles in the weight off obstacle clearance condition were significantly decreased compared to those in the weighted condition (Figure 34).

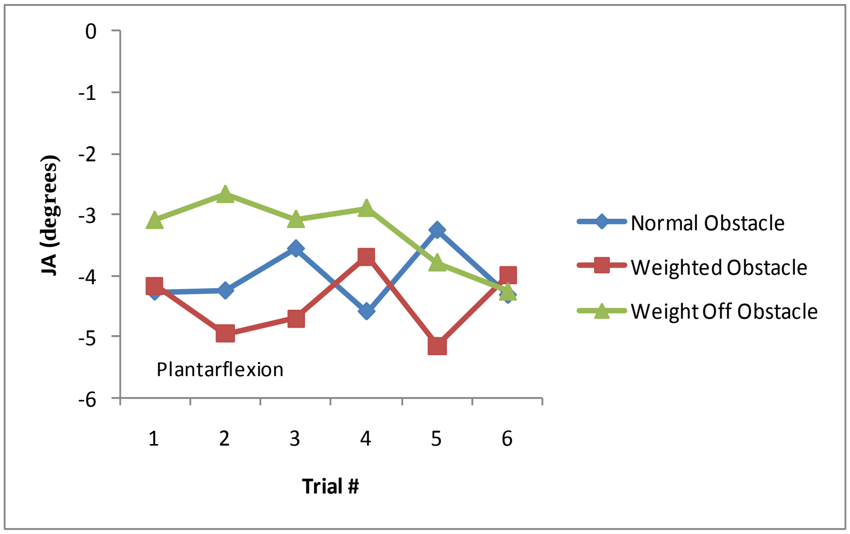


Figure 34: Ankle joint angles at heel contact (obstacle)

Another significant trial by condition interaction effect was detected (F=2.37, p=0.0153) for the hip joint angles at heel contact. Post hoc Tukey tests uncovered that in trials 3 through 6, the hip joint flexion angles in the weighted obstacle clearance condition were increased compared to those in the normal and weight off conditions (Figure 35).

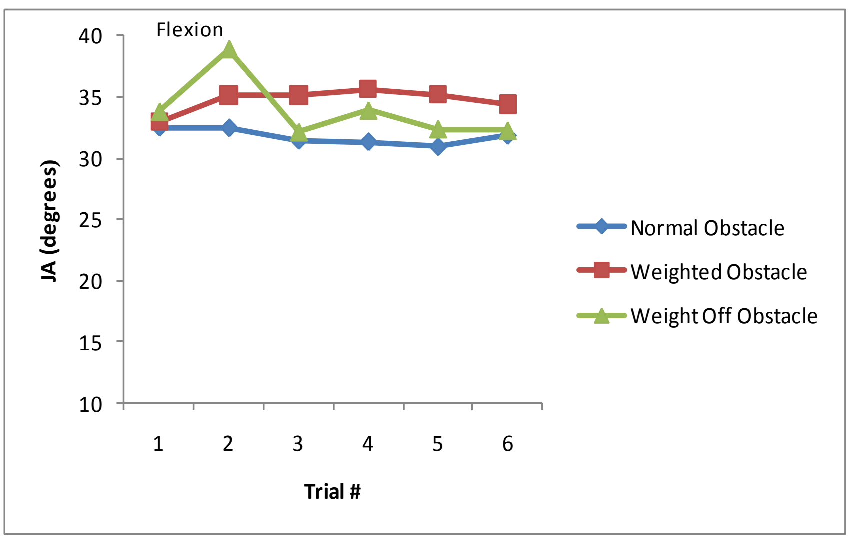


Figure 35: Hip joint angles at heel contact (obstacle)

The joint power data results at heel contact proved to be very similar to those found for the joint angle data. A significant trial by condition interaction effect (F=2.6, p=0.008) was found for the ankle joint power at heel contact. Post hoc Tukey tests performed on this significant heel contact result determined that, in trials 1 through 3, the weighted obstacle clearance condition produced ankle joint powers different from those in the normal obstacle clearance condition (Figure 36). No real pattern was established, but nonetheless the weighted condition was different than normal and changed from an energy absorption system to an energy generation system.

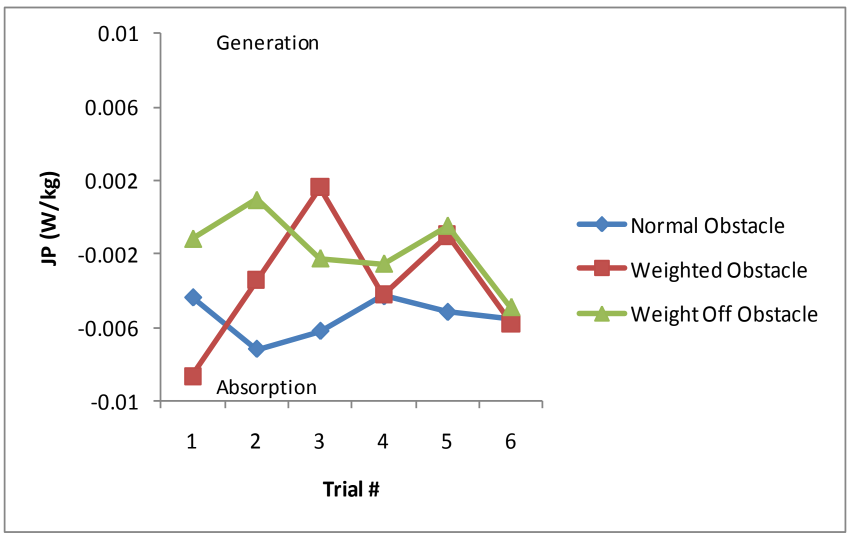


Figure 36: Ankle joint powers at heel contact (obstacle)

Another significant trial by condition interaction effect (F=2.11, p=0.0316) was detected for the hip joint power at heel contact. The post hoc Tukey tests performed showed that the hip joint power in trial 1 for the weighted obstacle clearance condition was increased, changing from an energy absorption to generation role, compared to the normal and weight off conditions (Figure 37). Differently from the ankle joint power, this effect was not persistent throughout further trials.

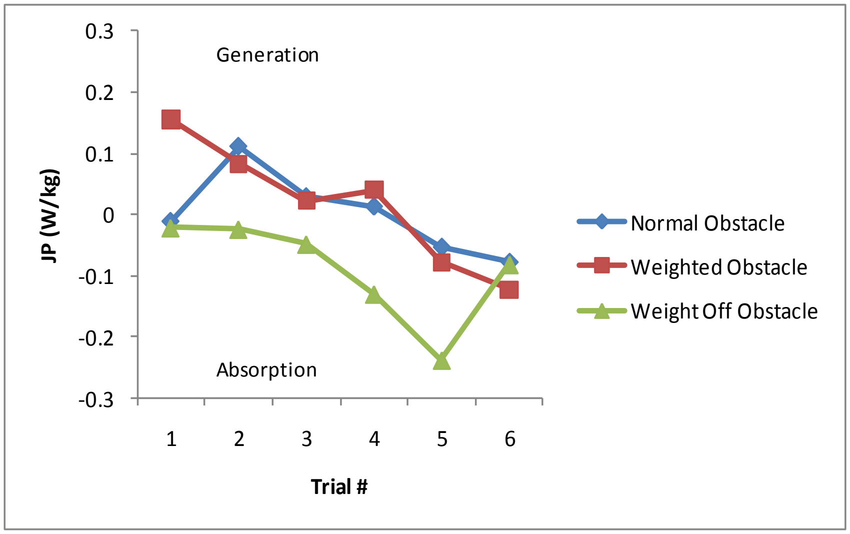


Figure 37: Hip joint powers at heel contact (obstacle)

No significant trial by condition interaction effects were detected by statistical analysis in both the horizontal direction and vertical direction for the right toe velocity variable at heel contact. However, a significant condition main effect was present in the horizontal direction (F=3.64, p=0.047) at heel contact. Specifically, it was determined that in the weight off obstacle clearance condition, the toe velocity was decreased compared to the normal condition.

Shown in Appendix D (viii) is a summary sheet of the statistical results detected at the ankle, knee and hip joints during obstacle clearance at heel contact.

4.13: Toe Obstacle Clearance

When running the repeated measures ANOVAs on the toe obstacle clearance variable, a significant trial by condition interaction effect (F=5.73, p=<0.0001) was detected. Performing the post hoc Tukey tests determined that, in trial 1, there were significant differences in all three of the obstacle clearance conditions (normal, weighted and weight off) with the weighted condition producing a spiked increase in toe clearance values (Figure 38). Then in trials 2 through 6, the normal and weighted obstacle clearance conditions seemed to level off to similar values while the weight off condition had toe clearances significantly decreased compared to them (Figure 38).

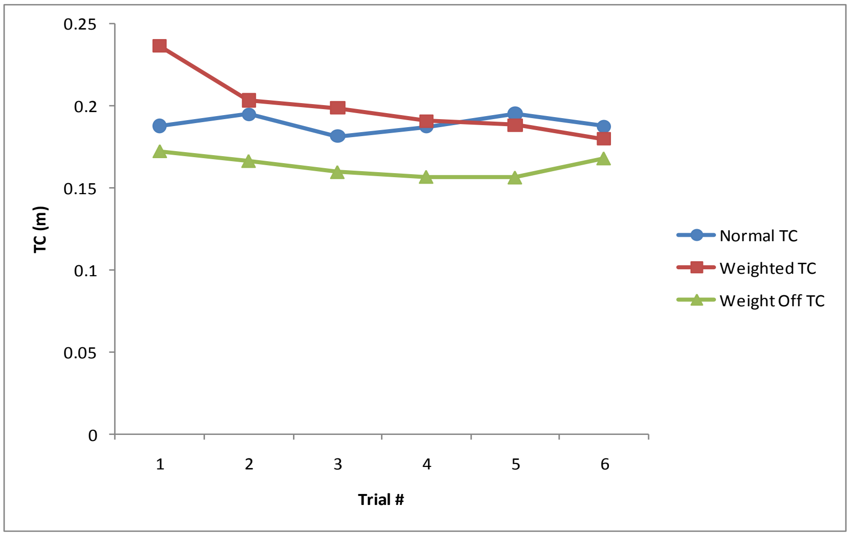


Figure 38: Toe obstacle clearance

4.14: Obstacle Toe Clearance versus Adapted Obstacle Toe Clearance

No surprises were found when looking at the toe obstacle clearance values from the normal obstacle clearance condition of the ten subjects in group 2 against the last twenty trials of the ten subjects in group 1 where obstacle clearance was done when already adapted to the limb weight. After performing statistical analysis, no significant trial by condition or main effects were detected for the toe obstacle clearance variable (Figure 39). This confirms the third hypothesis discussed earlier, that is, there was no difference between normal obstacle toe clearance and adapted obstacle toe clearance.

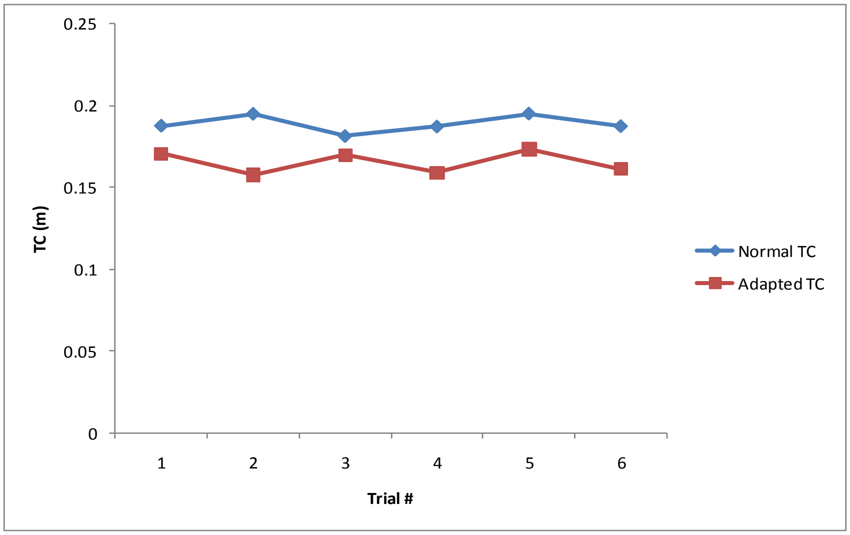


Figure 39: Obstacle toe clearance versus adapted obstacle toe clearance

Chapter 5

Discussion

5.0: Getting Started

Before sense can be made out of all these variables analyzed, the normal function of both level ground walking and obstacle clearance tasks needs to be more fully understood. Not only what is involved in each, but also how they are different and how the addition and removal of weight may affect them differently. During gait, the body must progress forward by generating mechanical energy while maintaining whole body balance with an upright posture, effectively control for adequate foot trajectory and adjust the energy usage at foot contact for proper shock absorption (Winter DA, 1991). In normal walking this process is quite automatic, but in a task such as obstacle clearance, this is not always the reality. In fact, the same basic walking functions just discussed in normal walking, still need to be accomplished in obstacle clearance but with the integration and addition of multiple voluntary modifications. More energy generation is needed, greater knee flexion often results causing an increase in the toe clearance or safety margin and shock absorption becomes even more important at heel contact following clearance due to the rapid lowering of the leg. This makes obstacle clearance a much more complex and volitional movement requiring more attention from multiple movement control centers in order to accomplish these simple normal walking functions. For this reason, both normal level ground walking and toe obstacle clearance tasks needed to be examined.

Additionally, it is also important to realize that some of the effects seen in the obstacle clearance group may not have been fully detected or as easily identified compared to those in the normal walking group. This is because, like stated above, obstacle clearance is a much more complex task. Therefore, because even the normal condition of obstacle clearance already incorporated exaggerated kinematics and kinetics, the limb movement patterns seen already contained a substantial built in safety margin. In other words, the added energy needs or compensation occurring due to the altered limb mechanics, may be masked, as the actual task of obstacle clearance itself results in variations that would be similar to that which is required in weighted movements. This could have potentially diminished the effects seen in joint angles, moments and powers when the weight was added and removed from the limb.

Studying both weight addition and removal also becomes very important in determining how the body adapts to a unilateral lower limb perturbation. It was both needed and expected that adding weight to the leg would cause changes. In some way or another, the body had to maintain its normal gait functions through adjustments in kinematics and kinetics. What becomes more interesting, is what happens in weight removal. The body seemed to return very quickly in some cases to normal kinematic and kinetic patterns while in other instances, a new steady state pattern was created. It appears that in weight removal, movement patterns do not simply return back to stored patterns, but instead, these patterns are adjusted and updated. Because of this, main effects detected show a change in movement patterns due to weight addition or removal alone, while interaction effects show an effect caused by the weight followed by an adaptation period back to normal or to a new steady state pattern.

5.1: Normal Walking Group

Progressing now in an order that examines joint angles, moments and powers at each joint separately, the results presented above will be more closely discussed. The ankle joint was affected at both toe off and during swing of the gait cycle. At toe off, the weighted and weight off conditions both had decreased ankle plantarflexion. This was especially prevalent in the first three trials of the weight off condition. Only the weighted condition produced effects during swing though where ankle dorsiflexion was increased. Both these results point to the weight forcing the ankle into having less effective extension with lasting effects still present when the weight was removed resulting in a new steady state movement pattern.

Moving to the knee next, significant effects were detected during stance, swing and at heel contact. During stance, the knee joint angle was altered where the weighted and weight off conditions had increased knee flexion angles compared to the normal walking condition. The peak during swing showed that the knee flexion angles in the first two trials decreased in the weighted condition compared to normal but increased compared to normal in the weight off condition. Interestingly, the added weight prevented the knee from flexing fully but was quickly adapted for because in trials 2 through 6, knee flexion angles leveled off to normal values. But, in the weight off condition, knee flexion angles remained increased compared to normal again showing this new steady state pattern. However, both the weighted and weight off conditions produced increased knee flexion angles at heel contact.

Lastly, the hip joint was found to be very effected by the addition and removal of a limb mass, as it showed significant interaction effects at all of the instances analyzed in the gait cycle (stance, TO, swing, HC). During stance and at toe off, both the weighted and weight off conditions produced decreased hip extension angles. Interestingly, this effect was more prominent in the weight off condition for trial 1 during stance and all six trials at toe off. Again, this result shows that the weighted and weight off conditions force the leg into having increased flexion with lasting effects still present when the weight was removed resulting in a new steady state movement pattern. During swing and at heel contact the hip joint flexion angles were increased in both the weighted and weight off conditions. Interestingly, this result was again, more prevalent in weight off condition across all six trials showing an adjustment to stored movement kinematic patterns resulting in a new steady state.

These results for the joint angles are quite similar to those found in previous research work. Generally, increased joint flexion angles were found at the joints more proximal to weight placement (Martin et al, 1990 and Reid and Prentice, 2001). In this current research, increased knee flexion was seen at the knee during stance, swing and at heel contact, except for the first two trials in the weighted condition during swing where the knee flexion angles actually decreased compared to normal. The results found at the hip, where significant effects were detected during stance, at toe off, during swing and at heel contact also followed what Martin et al, (1990) and Reid and Prentice, (2001) determined. That is, increased flexion at joints proximal to weight placement. Interestingly, even though hip flexion angles were increased in the weighted condition, this effect was even more prominent in the weight off condition. This suggests one of two possibilities. First, the body's internal model which has preprogrammed movement strategies is more affected by weight removal or an un-adaptation compared to weight addition resulting in a new steady state movement pattern seen in the weight off condition. Secondly, the experimental protocol may not have been long enough to allow for a complete recalibration of movement patterns back to normal. However, a return to normal patterns was not always found in Noble and Prentice, (2006) either. In any case, it seems that with a distally placed weight, the hip is the most important joint used to modulate the weighted limb trajectory throughout a gait stride because trial by interaction effects were found at all four of the discrete points analyzed (stance, toe off, swing and heel contact). The knee joint would be the next most important joint. It also seems though, that, differently from Martin et al, (1990) and Reid and Prentice, (2001), the ankle joint angle is utilized at toe off and during swing for further control of limb trajectory due to the fact that significant effects were detected there as well. A reason for this ankle joint angle increase at toe off and during swing may be due to the different placement of the limb weight on the shank. Most previous studies (Martin et al, 1990 and Smith et al, 2007) employed a method of attachment around the shank or thigh joint centre of mass while in the current study the limb weight was placed just proximal to the ankle maleoli reflective markers. Perhaps when the weight is placed as distally as possible to the ankle, more ankle control is relied upon for the control of the foot due to the larger inertial forces that the weight induces on the leg. This point however, will be revisited when joint moments and powers are discussed in upcoming sections.

Progressing onto the kinetics now and starting with the joint moments, the ankle joint moments, were also altered at toe off and during swing of the gait cycle. At toe off, in the weighted condition, the ankle joint plantarflexor moments were increased compared to those in the normal and weight off conditions. This effect was especially prevalent in trials 1 through 3 before leveling off in the remaining trials to a steady state. The ankle joint plantarflexor moment was also increased in the weighted condition during swing.

The knee joint moment seems to be a very important variable controlled for as significant effects were detected during stance, toe off, swing and at heel contact. During stance and at toe off, the knee joint extensor moment was increased in the weighted and weight off conditions compared to normal. However, this effect was most prominent in the weighted condition. Conversely, during swing the weighted condition caused the knee joint flexor moment to increase compared to both other conditions. The strongest knee joint moment change, however, occurred at heel contact. The knee joint showed that all trials and all conditions were different from each other with the joint flexor moments being decreased in the weighted condition and in the weight off condition the knee joint moment was actually changed into an extensor moment across all trials, showing again this formation of a new steady state pattern of movement.

Interestingly, the hip joint showed significant effects at all of the same points that the knee joint did. During stance and at toe off, the hip flexor moments were increased in the weighted and weight off conditions compared to normal. Again, this effect was more prominent in the weighted condition. Then, during swing the hip extensor moment was now increased in the weighted condition. Like the knee joint as well, the largest effect on hip joint moment occurred at heel contact. This time, the hip joint flexor moments decreased across all trials and all conditions. However, at the hip, the weighted condition, rather than the weight off condition, was decreased so much that the hip joint moment became an extensor moment.

A couple different points need to be addressed here. First, during stance and toe off, the ankle joint plantarflexor moments, knee joint extensor moments and hip joint flexor moments all increased in the weighted condition. These results were comparable to those of Martin et al, (1990) and Smith et al, (1997). This trade off taking place between the knee and the hip joints, suggests intercompensation occurring. In stance, changes to knee and hip joint moments are needed in order to accurately keep the support moment of the leg within appropriate and balanced standards (Winter DA, 1991). Furthermore, this intercompensation occurring between the knee and hip, also allows the body to maintain accurate control of the torso and centre of mass from stride to stride (Winter DA, 1991). Secondly, the results in swing were not quite as easy to interpret. The knee flexor and hip extensor moment increases during swing were very comparable to Martin et al, (1990) and Smith et al, (1997). However, in their experiments, changes were only found at the joints proximal to weight placement and no effect was found at the ankle during swing. In this current research, the weighted condition produced an increased ankle plantarflexor moment. Again this effect at the ankle could have possibly been due to where the limb weight was placed on the shank. Perhaps, like the ankle joint angle, when the limb weight was placed distally on the shank more control of foot trajectory was needed due to the larger inertial forces acting on the leg.

Another interesting result found was how at heel contact, the knee and hip joint flexor moments decreased. However, at the knee during the weight off condition, an extensor moment was produced and at the hip during the weighted condition, an extensor moment was produced. There seems to be a knee-hip trade off effect taking place where, with weight removal, the knee was more responsible for proper foot placement but with the addition of weight the hip now took over this role. This possibly, can be explained by the hip being utilized more robustly when the leg is heavier and increased energy demands are present while the knee is used more for better control of foot placement when the weight is removed.

Furthermore, it was expected that in the weight off conditions (after weighted), joint variables would return to values more similar to that of the normal condition (Smith et al, 1997 and Noble and Prentice, 2006). At heel contact for the knee and hip joint moment this effect did not occur. In fact, the weight off condition produced knee joint moments that changed from flexor to extensor and never returned back to normal. This could mean that, following weight removal, a whole new walking pattern was established or again that the experimental protocol did not allow enough time for the knee joint moments to return back to normal.

Continuing onto the joint powers, the ankle joint again, showed significant effects at toe off and during swing. In both cases, the ankle joint decreased energy generation in the weighted condition. In fact, at toe off, the trial 1 ankle joint power changed from an energy generating system to an energy absorbing system.

Other than in the stance phase, the knee joint had altered joint powers in all the other gait instances analyzed (TO, swing, HC). At toe off, knee joint energy absorption was increased in both the weighted and weight off conditions while during swing, this was still true but only in the weighted condition. The knee joint was also the only joint to show an altered joint power at heel contact. It showed that the weighted condition had increased knee joint energy generation (especially in trial 1) before leveling off to similar values seen in the normal and weight off conditions in later trials. Interestingly, the normal and weight off conditions remained as energy absorption systems throughout the experimental trials but in trials 1 through 5 (especially trial 1) the knee joint changed into an energy generating system before leveling off in trial 6. Meaning, that in order to control the leg and foot for proper landing when the leg was weighted, the knee took a more active role by increasing energy generation.

Unsurprisingly, the hip joint was also heavily affected by weight addition and removal. Evidence for this was seen during stance, at toe off and during swing. In fact, during stance, only the hip joint power showed a significant effect. Other than the weight off trial 1 result, the weighted and weight off conditions were quite similar to each other, with the important result being that the hip joint increased its energy generation output. This result implies that in stance, the hip needed to produce more power in order to propel the weighted right leg into swing. Again this result was still prevalent after weight removal and did not return to normal implying that a new kinetic steady state motor pattern was established. Alike, the joint powers at toe off and during swing also showed an increase in energy generation in the weighted and weight off conditions compared to normal.

In summary, it appears that the joint powers in the leg have a few key characteristics. First, the ankle joint decreases energy output. Secondly, other than at heel contact, the knee joint seemed to increase energy absorption throughout the gait cycle. Lastly, the hip joint increased energy generation at nearly all discrete points looked at in the gait cycle for both the weighted and weight off conditions.

These joint power results again show that the weighted condition generally increases joint powers and energy generation, whether it is for the first couple trials or throughout all trials of the test protocol (Martin et al, 1990). Of the three joints in the limb, the hip joint increased energy generation the most and at multiple instances in the gait cycle, meaning that the hip was most important for adjusting and compensating for the added limb mass. Although joints proximal to the limb weight were affected, the distal ankle joint power also showed significant decreases in energy generation at toe off and in swing. So, it seems that the control of the ankle joint angle, moment and power is a key feature in controlling foot trajectory and compensating for this altered mechanical perturbation that the added limb mass causes.

There are three reasons why this thesis research may have shown changes to gait mechanics distal to the limb weight placement, along with altered mechanics proximal to weight placement. First, it may have to do with where the limb weight was placed. Most previous studies have used a shank center of mass location for weight placement while this current research chose a more distally (just above the ankle) weight placement location. Because the weight was placed so close to the ankle joint, ankle joint control may have been utilized to a larger extent due to the larger inertial forces acting on the leg as a result of added mass. This may have decreased the demand placed on the knee and to a lesser degree the hip. Secondly, many controversies exist on whether treadmill walking and level ground walking produce similar kinematic and kinetic walking patterns. Studies such as those done by Riley et al, (2006) showed differences between 12 of 22 kinematic gait variables and 18 of 24 kinetic variables addressed, but because these differences were small, concluded that treadmill walking was similar to level ground walking. However, many other studies like Alton et al, (1998) showed larger differences in kinematic and kinetic variables of the normal gait pattern and concluded that significant differences do exist between treadmill and level ground walking strategies. In any case, it is sufficient to say that treadmill walking is at least minimally different from level ground walking and may explain why some of the results found in this thesis research showed changes in walking strategy distally to limb weight placement. Finally, an advantage to using treadmills is that when the limb weight is first applied, data collection can be continuous and all mechanical effects uninterrupted. With level ground walking, data collection cannot be continuous and there were breaks between trials. Even with rolling the subject back to the starting position on a chair, subjects are still aware of the added mass and may adapt to it in between trials due to proprioception alone. Therefore, results seen distally to limb weight placement could be a result of the larger knee and hip joints being manipulated more in between trials compared to the smaller ankle joint resting on the foot bar of the chair. Furthermore, overground walking effects the data collection within individual trials themselves. Participants took steps before and after the two force platforms; meaning that a number of steps in each trial were not analyzed.

These last two points addressed may also help explain why the results seen in this research are not as clean and clear cut as those seen in Noble and Prentice, (2006) and Smith et al, (2007). These two studies used treadmill collection techniques which allowed for continuous data collection, no time in between trials and analysis of multiple repeated strides at a time. For example, in Noble and Prentice, (2006), analysis was restricted to the first 250 strides in each experimental condition. In this current research collection was done overground which only allowed for about 4 strides per trial to be collected due to lab room size restraints. So a total of 80 strides per subject were possible to be analyzed in each experimental condition. Even more conflicting is again the time and proprioception taking place in between these trials.

Nevertheless, in this current research, adaptation to the added limb mass did occur in many of the variables analyzed in both the weighted and weight off conditions. In the weighted condition, these adaptations usually occurred immediately or in trials 1 through 3 resulting in evident increases or decreases in joint angles, moments and powers. After this immediate adaptation period, the joint angles, moments and powers either returned back to values similar to in the normal condition or leveled off, and established a new steady state. Although this effect was still seen in the weight off condition, interestingly, the weight off condition seemed to have a stronger effect compared to mass addition. In fact, for the variables of ankle plantarflexion in stance, knee flexion in swing, knee extensor moment at heel contact and numerous (5) hip joint variables analyzed, the weight off condition produced lasting effects. That is, for these variables completely new movement parameters were produced. In some cases, flexion angles became extension angles and flexor moments became extensor moments. It seems that with weight addition a major adaptation is required to accommodate the extra mass but this adaptation is quite brief. Whereas, with weight removal, an effect lasting at least 20 trials and producing totally new movement patterns become established, resulting in a much more prolonged adaptation period.

These results also show evidence of the human body utilizing an internal model of movement, even with a task as simple and automatic as normal walking. Hinder, MR and Milner, TE (2003) describe an internal model as a system that encodes muscle activation patterns while compensating for the effects of predictable environmental forces. Adding weight to the body is a much more familiar situation to participants in the real world compared to weight removal. This may explain why the effects seen in the weighted condition resulted in a shorter adaptation period. Participants were much more capable of predicting limb movement when the weight was attached then when it was removed. So, because this weight off condition was not as easily predicted for, joint variables in the weight off condition showed prolonged adaptation periods and new movement strategies. Furthermore, an internal model is formed by practicing the relationships between joint segments, velocities, torques and muscle activations where learning leads to its formation (Hinder MR and Milner TE, 2003). This fact also gives insight into what the current results are showing. That is, in the first few trials of each condition a learning process occurred where joint variables showed marked increases or decreases. Then, once the learning process was over, a new steady state resulted and an adjusted internal model for the appropriate movement was created.

Looking now at further discrete variables analyzed in the walking group (toe velocities and minimum toe clearance) other interesting results were found. At toe off, a significant effect was found for toe velocity in the vertical direction. Specifically, the weighted condition had decreased vertical direction toe velocities throughout all trials of the experimental protocol. If the experimental protocol would have allowed for more trials and longer collection intervals, perhaps this vertical toe velocity would have returned back to normal velocities, but in this case, a new steady state occurred like in Noble and Prentice, (2006) and Smith et al, (2007). Curiously though, at toe off, the weighted condition showed significant effects with the knee and hip joint moments and powers being increased. This means that the knee and hip joints were working harder at toe off but achieving a toe velocity in the vertical direction or pull off force that was still decreased compared to the normal and weight off conditions. Having a decreased toe velocity makes sense because of the added weight but it might also explain why the ankle joint angle, moment and power were increased in the weighted condition. Perhaps the knee and hip joints relied on the ankle joint for additional assistance in controlling foot trajectory through more fine motor control.

At heel contact, the weighted condition produced increased toe velocities in the horizontal and vertical directions. Although, not statistically significant, these effects were more pronounced in trial 1 before leveling off in trials 5 and 6. Unlike toe off, the ankle joint angle, moment and power did not seem to influence the foot at heel contact. Instead the proximal joints (knee and hip), like in Royer et al, (1995) and Martin et al, (1990), were the joints affected by showing significant effects.

When looking at the minimum toe clearance at mid swing of the walking group an interesting result occurred. A significant interaction effect was found showing that, in trial 1, the minimum toe clearance in the weight off condition spiked higher than the other two conditions before leveling off in trials 2 through 6. This is surprising because in most of the results so far, the weighted condition has produced results similar to this. It seems that the body's own representation of limb properties was more affected by weight removal than weight addition. So, because the body utilized the same limb movement strategies of the angles, moments and powers as when weighted, an increase in minimum toe clearance resulted but quickly returned back to normal standards in trials 2 through 6. However, in trials 3 through 6, the weighted condition did produce greater minimum toe clearances than the normal and weight off conditions. So, again the weighted condition still had a significant increase on the minimum toe clearance magnitude which suggests a safety precaution taken by subjects to ensure adequate foot trajectory throughout the swing phase of the gait cycle.

It should be noted, however, that the minimum toe clearance values for normal walking found in this research were slightly higher than those previously recorded in other studies. Mills et al, (2007) and Winter, (1992) recorded minimum toe clearances in the range of 0.5 – 2.5 centimeters. In this study, minimum toe clearances ranged from 5 – 7 centimeters. The reason for this was because the reflective toe marker was placed on the top of the shoe pointing upwards in the vertical direction, making a margin of error of about 2 – 3 centimeters from the actual toe. Other studies have placed the reflective markers in a position pointing medially outwards from the side of the toe. So if you subtract this margin of error from the minimum toe clearance values, the results obtained are quite comparable.

5.2: Obstacle Clearance Group

In the obstacle group, similar results to the walking group were found. Starting with the ankle, significant effects were detected at all four instances analyzed in the gait cycle. During the stance phase, the ankle showed increased dorsiflexion angles in the weighted condition throughout all experimental trials tested. Then, at toe off and during swing, the ankle joint plantarflexion angles were decreased in the weighted condition. However, the same was true at heel contact, but only in the weight off condition instead, for trials 1 through 3 compared to the weighted and normal obstacle clearance conditions. In the last three trials this effect was not apparent. Like Noble and Prentice, (2006) and Smith et al, (2007), the ankle joint angle at heel contact did show an adaptation effect where the first few trials showed a marked increase followed by a leveling off period before returning to a steady state.

The knee has been previously studied to show great changes in flexion when clearing an obstacle (Patla and Prentice, 1995). In the current results, increased flexion was detected for the knee joint during the stance and swing phases. During stance, this was due to the weighted condition. Oddly, at the peak in swing, knee flexion was increased considerably, but for almost all the experimental trials in the weight off condition.

As important as the knee joint is during the swing phase of obstacle clearance, the hip joint also plays a very large role. Evidence for this is seen by alterations to the hip joint angle at toe off, during swing and at heel contact. The weight off condition produced increased hip flexion angles at toe off and both conditions (weighted and weight off) produced increased hip flexion angles during swing. This was to a lesser degree in the weight off condition. At heel contact, the hip joint flexion angles were increased in the weighted obstacle condition in trials 3 through 6 compared to those in the normal and weight off conditions. This effect is however, different than the adaptation periods seen in Noble and Prentice, (2006) and Smith et al, (2007). It was unexpected that these changes would occur in trials later than the first couple.

Work done by Patla and Prentice, (1995) and Patla et al, (1993) showed that when traversing obstacles, increased flexion occurs at all three of the joints in the lower limb. In these current results, increased flexion occurred more substantially at the ankle and hip. In fact, throughout all the discrete variables in the gait cycle analyzed (stance, toe off, swing and heel contact), significant effects showed that the weighted condition produced increased joint flexion angles at the ankle and hip. Interestingly, the knee only showed increased flexion angles during the swing phase of obstacle clearance. However, Patla and Prentice, (1995) and Patla et al, (1993) only deemed the knee a significant contributor to obstacle clearance during swing as well. Either way, the current study still shows the same general trend of increased flexion throughout the gait cycle. On the other hand, those previous studies were testing different obstacle heights and did not involve added limb mass. These differences may also be due to the same reasons provided earlier in the normal walking group of weight placement, level ground walking versus treadmill walking and non-continuous data collection. Regardless, the current results show that during obstacle clearance with the addition and removal of limb mass, subjects chose a strategy of increased flexion at the ankle and hip with increased flexion at the knee during swing. This large role that the hip plays could also be related to other obstacle clearance strategies seen like "hip flexion" or "hip hiking" described by Lu et al, (2006) and Patla et al, (1993) respectively.

Changes in the joint angles in the weight off condition also show that the weighted condition produced lasting effects which may take longer to un-adapt for than this experimental protocol allowed. Decreases in ankle plantarflexion at heel strike, increases in knee flexion during swing and hip flexion at toe off and during swing in the weight off conditions suggest that, after weight removal, completely new obstacle clearance strategies or motor patterns were used to traverse obstacles. It could be that the movement pattern embedded in the body's internal model (Noble and Prentice, 2006) was changed, updated or even possibly improved by using a new steady state blueprint. This result again, like in the normal walking group, shows evidence for the existence of an internal dynamics models as stated by Hinder MR and Milner TE, (2003). Or it could also be that the experimental protocol of the research did not allow enough time for motor patterns to return to those used in normal conditions.

With the ankle joint showing so many statistical results, it was expected that the same would be present for joint moments as well. However, only the stance phase moment, resulted in a change at the ankle joint. The ankle plantarflexor moment was increased in the weighted condition.

The knee, now showing its more dominant role in obstacle clearance, produced significant results during stance, at toe off and during swing. The knee extensor moment increased during stance, then decreased at toe off and finally increased again during swing in the weighted condition. However, this decrease in knee extensor moment at toe off was primarily in the weight off condition and lasted only for trial 1. Like Noble and Prentice, (2006) and Smith et al, (2007), a quick adaptation period to weight removal was seen with marked changes occurring immediately followed by a leveling off period where knee joint moments returned to steady state.

Differently from the knee joint, the hip joint flexor moment during the stance and swing phases showed increases. Like the stance phase in the normal walking group, this increased knee extensor and hip flexor moment can be interpreted as a knee – hip trade off occurring. By controlling the support moment of the leg during the stance phase through intercompensation, trunk stability and centre of mass changes were controlled for from stride to stride (Winter DA, 2001).

Finishing off with the joint powers, the ankle joint power showed an increase in energy generation during stance for the weight off condition. At heel contact, the ankle joint powers in the weighted obstacle condition were different than those in the normal obstacle condition. In trial 1 the ankle joint energy absorption decreased compared to normal followed by a spiked increase in energy generation in trials 2 and 3 before leveling off to steady state values in the remaining trials of 4 through 6. It was as if there was an uncertainty period where the best strategy utilized by the ankle was being calibrated until finally becoming adapted to the added limb mass.

During stance, the knee joint increased power generation in the weight off condition. The most interesting result for the knee joint power occurred at toe off. As did the knee joint moment at toe off, the knee joint power at toe off also had a significant effect where, again, trial 1 in the weight off condition was different from the normal and weighted conditions. In trial 1 the energy absorption at the knee decreased and was very close to shifting into an energy generation role. Although, this result only lasted in trial 1 of the experimental protocol, a result that again is similar to the adaptation period and leveling off period seen in Noble and Prentice, (2006) and Smith et al, (2007). Additionally, Patla and Prentice, (1995) concluded that in obstacle clearance, the knee joint was actively controlled while the ankle and hip joints are passively controlled through intersegmental dynamics. This current result seen during toe off for the knee was comparable. Interestingly, the weight off condition is where most of these effects at stance and toe off were seen. Perhaps the body's internal model (Noble and Prentice, 2006) is very robust in adapting to new mechanical properties of the shank. However, it may be that when these adaptations occur, reverting back to normal is a more difficult and confusing situation resulting in a new steady state with new motor patterns shown in the weight off condition.

Lastly, the hip had similar results to the ankle joint in terms of power. Energy generation was increased in both the weighted and weight off conditions during stance. The hip joint, at heel contact, showed an increase in energy generation for trial 1 in the weighted condition compared to the normal and weight off conditions. This effect did not last in trials 2 through 5 though, as the hip joint powers returned back to steady state baseline measures. Again here, like results seen in Noble and Prentice, (2006) and Smith et al, (2007), a quick adaptation period was seen where the hip joint power was increased followed by a return to baseline values once the body's internal model incorporated the new added mass and adapted for it.

In any event, at heel contact the ankle and hip joint angles and powers were found to be the key variables in the obstacle group to show strategic changes in locomotor patterns. However, unlike at toe off or in Patla and Prentice, (1995), the knee did not actively control this process. Instead both the ankle and hip had increased work profiles. However, this may not be such a conflicting result because MacLellan and Patla, (2006) showed increased work profiles at the ankle and hip, along with the knee, during obstacle clearance tasks. Perhaps the added limb mass forced the leg to change its motor patterns and utilize the ankle and hip more for a safer landing following obstacle clearance.

Looking now at the last three discrete variables analyzed, some similar results of those in the walking group occurred. First the toe velocity in the vertical direction at toe off showed a significant effect. The toe velocities in trial one were different between all three conditions (normal, weighted and weight off) with the weighted condition being decreased and the weight off condition being increased. This decrease in toe velocity in the weighted condition also continued throughout trials 2 to 5. In trial 6 however, all three conditions leveled off to similar values in each. As in other variables discussed above, ankle angle and hip power, the vertical direction toe velocity showed an initial adaptation period where toe velocities were decreased in the weighted obstacle condition but returned to baseline measures by trial 6 (Noble and Prentice, 2006 and Smith et al, 2007). This result of decreased toe velocities in the weighted condition, like in the normal walking group, makes sense because there is added weight attached to the leg. At toe off though there was significant effects for the knee moment and knee power variables. Both were increased meaning that the knee and hip were actively trying to work harder in order to achieve similar toe velocities as normal.

At heel contact, the only significant statistical result found was a weak effect for the toe velocity in the horizontal direction. Toe velocities in the weight off obstacle condition were generally lower than those in the walking group. This showed that after the addition and removal of the added limb mass a new baseline was established for appropriate toe velocity in the horizontal direction at heel contact. It was surprising though, those stronger significant effects were not found for the toe velocities in both the horizontal and vertical directions. Decreased toe velocities at heel contact when weighted could have been interpreted as a safety mechanism used in order to help the foot land more softly. However, Patla et al, (1993) did not find strong evidence either for decreasing toe velocities in both the horizontal and vertical directions when looking at different obstacle heights and widths. So it seems that the toe velocity at heel contact may be a pre set parameter where, regardless of obstacle task or mechanical manipulation, is practically unchanged.

Finally, a significant effect was found for the obstacle toe clearance variable. A few different results occurred when analyzing this variable. First, in trial one, all three of the experimental conditions produced significant statistical results compared to each other. The weighted condition showed a marked increase in toe obstacle clearance compared to the normal obstacle condition while the weight off condition was decreased compared to the normal obstacle condition. This spiked increase in trial 1 of the weighted condition showed a voluntary gait modification used as a safety precaution by the subjects in order to safely traverse the obstacle and maintain an adequate margin of safety for toe obstacle clearance. Secondly, this increase in toe obstacle clearance in the weighted condition did not remain in trials 2 through 6 showing the pattern of a quick adaptation period followed by a recalibration of movement strategy back to normal steady state conditions. This again was similar to findings in Noble and Prentice, (2006) and Smith et al, (2007). Lastly, throughout trials 2 to 6 the weight off condition had toe obstacle clearance values decreased compared to the normal and weighted obstacle conditions. It seems that after weight addition and removal, a new toe obstacle clearance movement strategy was used by the body's internal model that allowed for a decreased toe obstacle clearance with similar power profiles. Like certain variables seen in the normal walking group, the toe clearance variable showed a quick adaptation to the mass addition but a completely different strategy taken in the weight off condition. Perhaps, the weight removal is not as easy a situation to predict compared to weight addition, resulting in more extensive changes needed to the body's internal model for that condition (Hinder MR and Milner TE, 2003).

Like the minimum toe clearance variable in the normal walking group, these toe obstacle clearance values may also be slightly larger than previously recorded in other studies. Again this is due to the reflective marker being attached to the toe pointing upwards in the vertical direction so there may be a margin of error around 2-3 centimeters. However, whether this margin of error is subtracted from the toe obstacle clearance results or not, the values for toe obstacle clearance obtained are comparable to those in studies such as Patla et al, (1993), Lu et al, (2006), Draganich et al, (2004) and MacLellan and Patla, (2006).

5.3: Normal Walking Group versus Obstacle Clearance Group

Most research studies do not compare level ground walking to obstacle clearance tasks. The reason for this being that the two movement tasks themselves are so different from each other. Level ground walking is a much simpler task that is very automatic to humans and often done without much attention required. The task itself takes use of intersegmental dynamics and energy transfers between different segments of the lower body. Obstacle clearance requires voluntary modifications to be integrated into normal gait patterns, increased muscle demand at all of the joints in the lower limb and more attention due to visual input used in scaling the object. So, because of these reasons stated above, it makes sense that in the current study more statistical differences were found in the normal walking group compared to the obstacle clearance group. Table 6 below shows this result and how much more affect the weight had on normal walking than obstacle clearance.

Table 6: Significant statistical results found between 2 experimental groups

	Normal Walking Group					Obstacle Clearance Group				
	Ankle	Knee	Hip	Other	Total	Ankle	Knee	Hip	Other	Total
# of Main Effects	3	7	5	2	17	5	4	5	1	15
# of Interaction Effects	3	3	6	2	14	3	3	2	2	10
# of Significant Post Hoc Tukey Tests	19	33	83	19	154	19	18	10	25	72
Joint Total	25	43	94	23		27	25	17	28	
Total					185					97

Another result that Table 6 shows is where the results occurred in each of the experimental groups. In the normal walking group many more results were found at the knee and especially the hip joint compared to the ankle joint. This was significant because these joints are proximal to the weight placement and agree well with previous work done by Martin et al, (1990), Reid and Prentice, (2001) and Royer et al, (2005). In the obstacle group, the results found were more evenly spread out across the ankle, knee and hip joints. Specifically, a significant adjustment was not located at any single joint but rather spread out between all three joints of the lower limb working together in order to accomplish the task. Previous studies by Patla et al, (1993), Maclellan et al, (2006) and Patla and Prentice, (1995) also showed that adjustments to all three of the joints in the lower limb had to be made in order to safely traverse obstacles.

Lastly, Table 6 also shows that the number of main effects and interaction effects found by statistical analysis in the normal walking and obstacle clearance group were generally equal. It was expected that like in studies by Noble and Prentice, (2006) and Smith et al, (2007) more trial by condition interaction effects would be present. That is the addition and removal of weight would produce altered gait characteristics over a short adaptation period followed by a recalibration back to normal or a new steady state pattern. However, this again may be due to the location of weight placement, treadmill versus level ground walking differences and time spent in between trials. In any case, the weight addition and removal conditions forced significant changes to the motor patterns used by the body's internal model which subjects chose to walk and safely traverse obstacles.

An alternative way to analyze the results seen above is to look at whether the majority of results seen were kinematic or kinetic. This is important because it can help determine what participants more prominently controlled for; joint angles and velocities (kinematics) or joint moments and powers (kinetics). Table 7 below shows the results seen in a kinematic versus kinetic representation.

Table 7: Kinematic and kinetic results found in both experimental groups

	Normal Walking Group		Obstacle Clearance Group	
	Kinematic	Kinetic	Kinematic	Kinetic
# of Main Effects	5	12	7	8
# of Interaction Effects	8	6	6	4
Total	13	18	13	12

In the walking group, more kinetic differences were detected compared to the kinematic variables. This result can be compared to Selles et al, (2004) where it was determined that mass perturbations to the lower leg of transtibial amputees resulted in a kinematic invariance strategy used by the subjects. This meant that, subjects made adjustments to joint kinetics more than the joint kinematics; a result which is comparable to the walking group in this research. Although the obstacle clearance group did not show this same effect, it can be argued again that obstacle clearance is a much more complicated task than walking so changes made by the participants were needed in both the kinematic and kinetic variables.

5.4: Obstacle Toe Clearance and Adapted Obstacle Toe Clearance

The ability of the body to use pre set motor patterns and then update these patterns to different situations is quite amazing. Looking at the normal obstacle clearance trials from the obstacle group against the adapted trials taken from subjects in the walking group, no differences in toe obstacle clearance were detected. This means that the body was very robust in adapting to the added limb weight. Once adapted, the same pre set motor pattern used with no added weight was still employed in order to maintain a sufficient toe obstacle clearance. It seems that toe obstacle clearance was a key variable monitored when traversing obstacles. Rather, changes in limb kinetics and kinematics were what was actively controlled for and increased when added weight was attached to the leg.

5.5: Results in the Real World and Future Research Directions

The results obtained in this thesis research have many theoretical implications related to them. Contributions to both the human movement and neuroscience fields of study have been demonstrated. Understanding how humans adjust to altered mechanical situations in their environment through changes in kinematics, kinetics and neural movement patterns is very important.

Other than the purely theoretical nature of this research, it also, has many implications for real world situations. Prosthetic engineering companies can use this thesis research in order to design a lower leg prosthetic that will allow for the most natural human movement. Changes in segment weight, segment moment of inertia and limb properties are all important factors that go into the design of a prosthetic limb. Knowing how the limb adjusts and adapts to a weight at a certain location can give insight into where the bulk of a prosthetic mass should be located. As well, knowing what joints become altered the most and what joints need to increase energy demands, will help guide the manufacturing design. Furthermore, diseases like Parkinson's, conditions like a stroke, sports injuries and industrial injuries usually only effect one leg of the body. Based on the results determined in this research thesis, rehabilitation programs and exercises can be developed to help strengthen the affected limb. Noticeable one sided impairments can help to be decreased by restoring the bilateral symmetry of the lower limbs and therefore, decreasing the risk of a fall. Although, this research is only a small part of the big picture and much more research into these areas of study is needed.

In biomechanics research, a problem that may arise is deciding exactly what variables to analyze due to the large number of options available. In the future, it would be very beneficial to analyze some other variables with this data that has already been collected. For example, looking at the changes seen in the amount of time spent in the stance and swing phases of the gait cycle may provide supplementary information into the adaptations occurring with the addition and removal of a limb weight. Or, perhaps looking into more detailed variables at the pelvis like pelvic tilt and obliquity and at the trunk like trunk roll may provide insight into other specific strategies participants used in order to accommodate to the changed mechanical parameters of the right shank. Specific to the obstacle clearance group, looking into the trail limb toe obstacle clearance, kinetics and kinematics may also help better understand how changing mechanical properties effects the symmetry of the two legs.

Furthermore, if this type of limb weight addition, removal and adaptation research were to be conducted again in the future during overground walking, two limitations need to be addressed. First, a better way to move the participant back to the starting position in between trials is needed. One idea may be a wheelchair where the participant's feet are completely supported throughout transport. Secondly, to even eliminate this problem above completely, a way to collect this data in one large trial having no time in between trials would be best. This would allow for no adaptation or proprioception occurring in between trials when no data was being collected.

Chapter 6

Conclusions

6.0: Normal Walking Group

This study sought to determine if previous strategies and adaptations caused by the addition and removal of a lower limb weight existed when using level ground force platform collection techniques. With reference to joint angles, the hip joint seemed to be the primary part utilized by the body's internal model in order to control for the weight addition and removal. At the hip, significant changes were found at all the instances analyzed in the gait cycle; stance, toe off, swing and heel contact. These typically occurred within the first one or two trials of the experimental conditions showing a quick adaptation period before settling back to normal or a new steady state and were interestingly, more prevalent in the weight off condition. The joint moments analyzed in the gait cycle showed that both the hip and knee were closely monitored in order to control for weight addition and removal. During stance and at toe off the knee extensor moments were increased while the hip flexor moments were increased. During swing the knee flexor moments and hip extensor moments were increased. These effects were more prominent in the weighted condition and suggest a knee-hip trade off; indicating very coordinated intersegmental dynamics. When looking at the joint powers another identifiable trend was detected. Other than at heel contact, the joint powers at the ankle and knee either decreased energy generation or increased energy absorption while the hip increased energy generation. This effect was seen in both the weighted and weight off conditions showing again this knee-hip trade off effect with highly coordinated intersegmental dynamics.

6.1: Obstacle Clearance Group

Although the obstacle clearance group was not as strongly affected by the weight addition and removal compared to the walking group, results were still identified. Interestingly, the ankle joint was the location where significant kinematic results were detected at all the instances analyzed in the gait cycle. Significant changes were found during stance, at toe off, during swing and at heel contact due to the weight condition. At these instances, the ankle was always plantarflexed less or dorsiflexion was increased. Like the normal walking group, the hip joint also had increased flexion during the weighted and weight off conditions at nearly all instances during the gait cycle. These results typically occurred within the first one or two trials of the experimental conditions showing a quick adaptation period before settling back to normal or a new steady state and were more prevalent in the weighted condition. The joint moments also showed two trends throughout the gait cycle. First during stance, at toe off and in swing knee extensor moments increased in the weighted condition but decreased in the weight off condition. Secondly, the hip flexor moments at these same instances increased (more so in the weighted condition). Both these results suggest, again, a knee-hip trade off but to a lesser degree than what was present in the walking group. Looking at the joint powers, the most identifiable pattern established by participants was found during stance and at heel contact. At these instances energy generation increased at both the ankle and hip joints. This effect was more prominent in the weighted condition and was especially true for early trials in the experimental conditions showing again this quick adaptation period before leveling off to normal or a new steady state in later trials.

Another important variable analyzed in the obstacle clearance group was toe obstacle clearance. The weighted condition toe clearance spiked in trial one but leveled off to normal toe clearance values in later trials. This illustrates both the quick adaptation period seen in other variables and a voluntary gait modification to the mass addition that can be interpreted as a safety precaution taken in order to assure that the obstacle was not contacted. During weight removal, toe clearance parameters were decreased and never returned to normal suggesting that a new obstacle clearance strategy developed.

6.2: Global Conclusions

Considering both the walking group and obstacle clearance group together it is adequate to say that the addition and removal of a lower limb mass does affect the kinematics and kinetics of human movement patterns. In many of the variables analyzed a very short adaptation period occurred in the first few trials of the weighted conditions followed by a progression back to baseline measures or a new steady state over the final trials. However, the results obtained are not quite as clean cut as what was found by Noble et al, (2006) or Smith et al, (2007). This is most likely due to the differences in weight placement location and a result of using level ground collection techniques rather than a treadmill. Treadmills allow for continuous collection, no time in between trials and a greater amount of strides to be taken by the participant. Nevertheless, the human body seems to be very robust in adapting to unilateral mass perturbations by updating the body’s internal model of movement patterns very quickly or even changing it so that improved patterns can be utilized.

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Appendix A: Right Leg Dominance Questionnaire

Name: _________________________________ Age: ______ Sex: M F

Instructions: Answer each of the following questions as best you can. If you always use one foot to perform the described activity, circle **Ra** or **La** (for right always or left always). If you usually use one foot circle **Ru** or **Lu,** as appropriate. If you use both feet equally often, circle **Eq**. Please do not simply circle one answer for all questions, but imagine yourself performing each activity in turn, and then mark the appropriate answer. If necessary, stop and pantomime the activity.

1. Which foot would you use to kick a stationary ball at a target straight in front of you?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
2. If you had to stand on one foot, which foot would it be?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
3. Which foot would you use to smooth sand at the beach?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
4. If you had to step up onto a chair, which foot would you place on the chair first?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
5. Which foot would you use to stomp on a fast moving bug?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
6. If you were to balance on one foot on a railway track, which foot would you use?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
7. If you wanted to pick up a marble with your toes, which foot would you use?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
8. If you had to hop on one foot, which foot would you use?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
9. Which foot would you use to help push a shovel into the ground?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
10. During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first?	**La**	**Lu**	**Eq**	**Ru**	**Ra**
11. Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities?		**YES**		**NO**	
12. Have you ever been given special training or encouragement to use a particular foot for certain activities?		**YES**		**NO**	
13. If you have answered YES for either question 11 or 12, please explain: (use back if necessary)					

Appendix B: Statistical Proof for Binning Trials 6-20 Together

Trials 6 to 20 not binned together (p-values)				
(1) Normal Walking Group				
	Stance	Toe Off	Swing	Heel Contact
Ankle Angle	0.9243	0.4017	<0.0001	0.7613
Ankle Moment	0.9584	0.0877	0.6991	0.5415
Ankle Power	0.4088	0.0736	0.1123	0.3911
Knee Angle	0.4701	0.7948	0.431	0.683
Knee Moment	0.8786	0.9093	0.1398	0.5228
Knee Power	0.0874	0.9972	0.3521	0.5213
Hip Angle	0.1024	0.3016	0.2981	0.4361
Hip Moment	0.2617	0.6908	0.3642	0.5883
Hip Power	0.7036	0.9607	0.3304	0.5557
Toe Velocity X		0.8687		0.5391
Toe Velocity Z		0.2834		0.4668
Minimum Toe Clearance		0.2003		
Trials 6 to 20 not binned together (p-values)				
(2) Obstacle Clearance Group				
	Stance	Toe Off	Swing	Heel Contact
Ankle Angle	0.5973	0.2531	0.5817	0.0916
Ankle Moment	0.8304	0.15	0.3364	0.0116
Ankle Power	0.6899	0.1357	0.4696	0.3884
Knee Angle	0.5201	0.6237	0.4896	0.0729
Knee Moment	0.5279	0.7756	0.4749	0.5219
Knee Power	0.0856	0.7127	0.4637	0.1846
Hip Angle	0.0327	0.6037	0.6722	0.1802
Hip Moment	0.2331	0.6912	0.1683	0.0706
Hip Power	0.4397	0.7563	0.1482	0.5388
Toe Velocity X		0.9205		0.2351
Toe Velocity Z		0.0827		0.2008
Minimum Toe Clearance		0.3619		
Adapted Obstacle Clearance		0.7974		

Appendix C: Example of Post Hoc Tukey Tests Done

Ex. Obstacle Toe Clearance

If Trial by Condition interaction effect was <0.05, then:

	N 1	Wt 1	WtOff 1	N 2	Wt 2	WtOff 2	N 3	Wt 3	WtOff 3	N 4	Wt 4	WtOff 4	N 5	Wt 5	WtOff 5	N 6	Wt 6	WtOff 6
N 1		<0.0001	0.046	0.2014	0.0286	0.0845	0.8894	0.1228	0.0157	0.6306	0.2811	0.0068	0.1865	0.6782	0.006	0.4531	0.6852	0.0262
Wt 1	<0.0001		<0.0001	<0.0001	0.0017	<0.0001	<0.0001	0.0002	<0.0001	<0.0001	<0.0001	<0.0001	<0.0001	<0.0001	<0.0001	<0.0001	<0.0001	<0.0001
WtOff 1	0.046	<0.0001		0.1911	0.0266	0.0899	0.9132	0.1158	0.017	0.6095	0.268	0.0074	0.1768	0.6564	0.0066	0.4288	0.7157	0.029
N 2	0.2014	<0.0002	0.1911		0.3511	0.0032	0.1571	0.7871	0.0003	0.4231	0.8398	0.0001	0.9649	0.3861	<0.0001	0.3157	0.0327	0.0001
Wt 2	0.0286	0.0017	0.0266	0.3511		0.0001	0.0202	0.5068	<0.0001	0.0849	0.2573	<0.0001	0.3741	0.0739	<0.0001	0.0242	0.0008	<0.0001
WtOff 2	0.0845	<0.0001	0.0899	0.0032	0.0001		0.112	0.0014	0.4744	0.0285	0.0058	0.3071	0.0028	0.0334	0.2882	0.0023	0.0505	0.8983
N 3	0.8894	<0.0001	0.9132	0.1571	0.0202	0.112		0.0931	0.0224	0.5355	0.2242	0.01	0.1448	0.5797	0.0089	0.3474	0.8297	0.0414
Wt 3	0.1228	0.0002	0.1158	0.7871	0.5068	0.0014	0.0931		0.0001	0.285	0.637	<0.0001	0.8212	0.2566	<0.0001	0.1711	0.0128	<0.0001
WtOff 3	0.0157	<0.0001	0.017	0.0003	<0.0001	0.4744	0.0224	0.0001		0.0041	0.0006	0.7581	0.0003	0.005	0.7271	<0.0001	0.0039	0.2692
N 4	0.6306	<0.0001	0.6095	0.4231	0.0849	0.0285	0.5355	0.285	0.0041		0.5487	0.0016	0.3983	0.9474	0.0014	0.9261	0.2887	0.0044
Wt 4	0.2811	<0.0001	0.268	0.8398	0.2573	0.0058	0.2242	0.637	0.0006	0.5487		0.0002	0.8056	0.5057	0.0002	0.4665	0.0618	0.0003
WtOff 4	0.0068	<0.0001	0.0074	0.0001	<0.0001	0.3071	0.01	<0.0001	0.7581	0.0016	0.0002		<0.0001	0.002	0.9672	<0.0001	0.001	0.1284
N 5	0.1865	<0.0001	0.1768	0.9649	0.3741	0.0028	0.1448	0.8212	0.0003	0.3983	0.8056	<0.0001		0.3626	<0.0001	0.2878	0.0283	<0.0001
Wt 5	0.6782	<0.0001	0.6564	0.3861	0.0739	0.0334	0.5797	0.2566	0.005	0.9474	0.5057	0.002	0.3626		0.0017	0.8547	0.3314	0.0057
WtOff 5	0.006	<0.0001	0.0066	<0.0001	<0.0001	0.2882	0.0089	<0.0001	0.7271	0.0014	0.0002	0.9672	<0.0001	0.0017		<0.0001	0.0009	0.1151
N 6	0.4531	<0.0001	0.4288	0.3157	0.0242	0.0023	0.3474	0.1711	<0.0001	0.9261	0.4665	<0.0001	0.2878	0.8547	<0.0001		0.0015	<0.0001
Wt 6	0.6852	<0.0001	0.7157	0.0327	0.0008	0.0505	0.8297	0.0128	0.0039	0.2887	0.0618	0.001	0.0283	0.3314	0.0009	0.0015		<0.0001
WtOff 6	0.0262	<0.0001	0.029	0.0001	<0.0001	0.8983	0.0414	<0.0001	0.2692	0.0044	0.0003	0.1284	<0.0001	0.0057	0.1151	<0.0001	<0.0001	

N = normal obstacle, Wt = weighted obstacle, WtOff = weight off obstacle

Within Trial post hoc Tukey tests

Appendix D: Post Hoc Tukey Test Results from Repeated Measures ANOVAs

i. Stance peak statistical results summary sheet

		Joint Angles			Joint Moments			Joint Powers		
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip
Trial	Interaction Effect									
	F-value			2.29						3.32
	p-value			0.0191						0.001
	Post Hoc Tukey Tests									
1	normal-weight			0.0141						<0.0001
	normal-weightoff			<0.0001						<0.0001
	weight-weightoff			0.0025						0.004
2	normal-weight			0.0063						<0.0001
	normal-weightoff			<0.0001						<0.0001
	weight-weightoff									
3	normal-weight			0.0003						<0.0001
	normal-weightoff			<0.0001						<0.0001
	weight-weightoff									
4	normal-weight			<0.0001						<0.0001
	normal-weightoff			<0.0001						<0.0001
	weight-weightoff									
5	normal-weight			<0.0001						<0.0001
	normal-weightoff			<0.0001						<0.0001
	weight-weightoff									
6	normal-weight			<0.0001						<0.0001
	normal-weightoff			<0.0001						<0.0001
	weight-weightoff									
	Main Effects									
	Condition Effect									
	F-value		11			3.85	21.74			
	p-value		0.0008			0.0407	<0.0001			

ii. Toe off statistical results summary sheet

		Joint Angles			Joint Moments			Joint Powers			Toe Velocities	
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip	X	Z
Trial	Interaction Effect											
	F-value	2.16		4.15	2.27			2.03				2.08
	p-value	0.0272		<0.0001	0.0203			0.0393				0.0338
	Post Hoc Tukey Tests											
1	normal-weight				0.0013			0.0397				0.0213
	normal-weightoff	0.0417		<0.0001								
	weight-weightoff				0.0005			0.0361				<0.0001
2	normal-weight											<0.0001
	normal-weightoff	0.0158		0.0011				0.0242				
	weight-weightoff				0.0001							<0.0001
3	normal-weight	0.0115		0.0023	0.0008							<0.0001
	normal-weightoff	0.013		<0.0001								
	weight-weightoff				0.0344							<0.0001
4	normal-weight			0.0216	0.0064							<0.0001
	normal-weightoff			<0.0001				0.0368				
	weight-weightoff											<0.0001
5	normal-weight			0.0095								<0.0001
	normal-weightoff			0.0008								
	weight-weightoff											<0.0001
6	normal-weight	<0.0001		<0.0001	<0.0001							<0.0001
	normal-weightoff	<0.0001		<0.0001								
	weight-weightoff				<0.0001			0.0446				<0.0001
	Main Effects											
	Condition Effect											
	F-value					11.46	13.66		6.8	8.74		
	p-value					0.0006	0.0002		0.0063	0.0022		

iii. Swing peak statistical results summary sheet

		Joint Angles			Joint Moments			Joint Powers		
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip
Trial	Interaction Effect									
	F-value		5.12	3.56						
	p-value		<0.0001	0.0005						
	Post Hoc Tukey Tests									
1	normal-weight		0.0235	0.0349						
	normal-weightoff		<0.0001	<0.0001						
	weight-weightoff		<0.0001	<0.0001						
2	normal-weight		0.0333	0.0427						
	normal-weightoff		<0.0001	<0.0001						
	weight-weightoff		<0.0001	<0.0001						
3	normal-weight			0.0015						
	normal-weightoff		<0.0001	<0.0002						
	weight-weightoff		<0.0001	0.0632						
4	normal-weight			0.0498						
	normal-weightoff		0.0005	<0.0001						
	weight-weightoff		<0.0001	0.0027						
5	normal-weight			0.0144						
	normal-weightoff		0.0004	<0.0001						
	weight-weightoff		<0.0001	0.0061						
6	normal-weight			<0.0001						
	normal-weightoff		<0.0001	<0.0001						
	weight-weightoff		<0.0001	<0.0001						
	Main Effects									
	Condition Effect									
	F-value	14.43			3.81	8.83	7.59	5.63	4.55	11.82
	p-value	0.0002			0.0418	0.0021	0.0041	0.0126	0.0251	0.0005

iv. Heel contact statistical results summary sheet of all variables

		Joint Angles			Joint Moments			Joint Powers			Toe Velocities	
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip	X	Z
Trial	Interaction Effect											
	F-value			3.19		2.34	3.78		1.98			
	p-value			0.0015		0.0165	0.0003		0.0447			
	Post Hoc Tukey Tests											
1	normal-weight					<0.0001	<0.0001					
	normal-weightoff			<0.0001		<0.0001	<0.0001					
	weight-weightoff					<0.0001	<0.0001		0.0041			
2	normal-weight			0.001		<0.0001	<0.0001					
	normal-weightoff			<0.0001		<0.0001	<0.0001					
	weight-weightoff					0.0004	<0.0001					
3	normal-weight			0.0001		<0.0001	<0.0001					
	normal-weightoff			<0.0001		<0.0001	<0.0001					
	weight-weightoff					0.0053	<0.0001					
4	normal-weight			<0.0001		<0.0001	<0.0001					
	normal-weightoff			<0.0001		<0.0001	<0.0001					
	weight-weightoff					0.0178	0.0001					
5	normal-weight			<0.0001		<0.0001	<0.0001					
	normal-weightoff			<0.0001		<0.0001	<0.0001					
	weight-weightoff					0.0381	0.0001					
6	normal-weight			<0.0001		<0.0001	<0.0001					
	normal-weightoff			<0.0001		<0.0001	<0.0001					
	weight-weightoff					<0.0001	<0.0001					
	Main Effects											
	Condition Effect											
	F-value		8.63								11.42	3.61
	p-value		0.0024								0.0006	0.0479

v. Stance peak statistical results summary sheet (obstacle)

		Joint Angles			Joint Moments			Joint Powers		
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip
Trial	Interaction Effect									
	F-value	5.75								
	p-value	<0.0001								
	Post Hoc Tukey Tests									
1	normal-weight	<0.0001								
	normal-weightoff									
	weight-weightoff	<0.0001								
2	normal-weight	<0.0001								
	normal-weightoff									
	weight-weightoff	<0.0001								
3	normal-weight	0.003								
	normal-weightoff									
	weight-weightoff	0.0014								
4	normal-weight	<0.0001								
	normal-weightoff									
	weight-weightoff	0.0002								
5	normal-weight	<0.0001								
	normal-weightoff									
	weight-weightoff	0.0003								
6	normal-weight	<0.0001								
	normal-weightoff									
	weight-weightoff	<0.0001								
	Main Effects									
	Condition Effect									
	F-value		5.88		15.79	11.02	6.71	5.84	18.25	6.53
	p-value		0.0108		0.0001	0.0008	0.0067	0.0376	<0.0001	0.0074

vi. Toe off statistical results summary sheet (obstacle)

		Joint Angles			Joint Moments			Joint Powers			Toe Velocities	
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip	X	Z
Trial	Interaction Effect											
	F-value					2.65			2.71			7.72
	p-value					0.0071			0.0059			<0.0001
	Post Hoc Tukey Tests											
1	normal-weight											0.0008
	normal-weightoff					0.0234			0.0332			<0.0001
	weight-weightoff					0.0019			0.009			<0.0001
2	normal-weight											<0.0001
	normal-weightoff											
	weight-weightoff											<0.0001
3	normal-weight											0.0017
	normal-weightoff											
	weight-weightoff											<0.0001
4	normal-weight											0.0013
	normal-weightoff					0.0197						
	weight-weightoff					0.0326						<0.0001
5	normal-weight											0.0068
	normal-weightoff											
	weight-weightoff											<0.0001
6	normal-weight											
	normal-weightoff											
	weight-weightoff											
	Main Effects											
	Condition Effect											
	F-value	6.19		7.26								
	p-value	0.009		0.0049								

vii. Swing peak statistical results summary sheet (obstacle)

		Joint Angles			Joint Moments			Joint Powers		
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip
Trial	Interaction Effect									
	F-value		2.58							
	p-value		0.0085							
	Post Hoc Tukey Tests									
1	normal-weight									
	normal-weightoff		0.0008							
	weight-weightoff		0.0071							
2	normal-weight									
	normal-weightoff		0.0208							
	weight-weightoff		0.0184							
3	normal-weight									
	normal-weightoff		0.0075							
	weight-weightoff									
4	normal-weight									
	normal-weightoff		0.0069							
	weight-weightoff									
5	normal-weight									
	normal-weightoff		0.0133							
	weight-weightoff		0.0106							
6	normal-weight									
	normal-weightoff		0.0021							
	weight-weightoff		<0.0001							
	Main Effects									
	Condition Effect									
	F-value	7.34		13.06		18.7	11.46			
	p-value	0.029		0.0003		<0.0001	0.0006			

viii. Heel contact statistical results summary sheet (obstacle)

		Joint Angles			Joint Moments			Joint Powers			Toe Velocities	
		Ankle	Knee	Hip	Ankle	Knee	Hip	Ankle	Knee	Hip	X	Z
Trial	Interaction Effect											
	F-value	1.97		2.37				2.6		2.11		
	p-value	0.0464		0.0153				0.008		0.0316		
	Post Hoc Tukey Tests											
1	normal-weight							0.0009		0.0462		
	normal-weightoff											
	weight-weightoff	0.047								0.0344		
2	normal-weight							0.0335				
	normal-weightoff											
	weight-weightoff	0.016										
3	normal-weight			0.0222				0.0413				
	normal-weightoff											
	weight-weightoff	0.034		0.041								
4	normal-weight			0.0086								
	normal-weightoff											
	weight-weightoff			0.0212								
5	normal-weight			0.0092								
	normal-weightoff											
	weight-weightoff			0.0433								
6	normal-weight			<0.0001								
	normal-weightoff											
	weight-weightoff			<0.0001								
	Main Effects											
	Condition Effect											
	F-value										3.64	
	p-value										0.047	