Role of the Lumbopelvic-Hip Complex in Bipedal Acceleration

by

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ABSTRACT

A majority of sporting competition involves short bursts of acceleration; yet most running literature focuses on constant velocity locomotion. The acceleration phase is a critical component to performance in athletes and is marked by increasing velocity, step lengths and arm and leg movements. However the purpose of the arm and leg motion is mostly speculative. Therefore the purpose of this project was to investigate the influence of arm and leg constraint on bipedal acceleration. Specifically, the goals of this study were: 1) To determine the role of the lumbopelvic-hip complex during the initial acceleration phase of a sprint, 2) To investigate the effects of constraint conditions of the upper arm (CUA), full arm (CFA) and hip constraint (CH) on the spatial-temporal kinematics during acceleration, 3) To investigate the effects of conditions CUA, CFA and CH on the joint kinematics during acceleration, 4) To investigate the effects of conditions CUA, CFA and CH on the gait kinetics during acceleration and 5) To investigate the effects of conditions CUA, CFA and CH on the latissimus dorsi (LD) and gluteus maximus (GM) muscle activity during bipedal acceleration.

The results demonstrate that the CUA condition decreased step length (SL_L), velocity (VEL), and anterior pelvic girdle rotation at initial contact ($APPGR_{IC}$). In contrast, an increase in stance time (ST_L) was noted with the CUA compared to an unconstrained (N) condition. The CFA condition produced decreases in SL_L, VEL, and

lateral pelvic girdle rotation at initial contact ($LPGR_{IC}$) while increasing ST_L. Furthermore, the CH condition decreased peak braking ground reaction force ($BGRF_{PEAK}$) and impulse (BGRI), SL_L, VEL, knee angle at toe off (KA_{TO}), $APPGR_{IC}$ and an increase in ST_L and ipsilateral LD activity at toe off ($ILLD_{TO}$). Results suggest primarily that arm constraints diminished acceleration. Additionally, the increase in arm motion in the presence of hip constraint suggests that the arms do more than conserve angular momentum within the LPHC. A cross effect was noted between the LD and GM while the LD was not significantly less active in arm constraints suggest that the LD has an additional role to the pelvis.

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List of Abbreviations

AA – Ankle Angle

APPGR – Anterior Posterior Pelvic Girdle Rotation

$BGRF_{PEAK}$ – Peak Braking Ground Reaction Force

BGRI – Braking Ground Reaction Impulse

CFA – Constrained Full Arm Condition

CH – Constrained Hip Condition

CL - Contralateral

CLGM – Contralateral Gluteus Maximus

CLLD – Contralateral Latissimus Dorsi

CUA – Constrained Upper Arm Conditon

EA – Elbow Angle

GM – Gluteus Maximus

LD – Latissimus Dorsi

HA – Hip Angle

IC – Initial Contact

IL - Ipsilateral

ILGM – Ipsilateral Gluteus Maximus

ILLD – Ipsilateral Latissimus Dorsi

KA – Knee Angle

LPGR – Lateral Pelvic Girdle Rotation

LPHC- Lumbopelvic-Hip Complex

SL – Step Length

LTF – Lateral Trunk Flexion

MVIC – Maximal Volitional Isometric Contraction

N – Normal Condition

NST – Non-Support Time

$PGRF_{PEAK}$ – Peak Propulsive Ground Reaction Force

PGRI – Propulsive Ground Reaction Impulse

SA – Shoulder Angle

sEMG – Surface Electromyography

ST – Stance Time

TF – Trunk Flexion

TLF – Thoracolumbar Fascia

TO – Toe Off

TR – Trunk Rotation

TVPGR – Transverse Pelvic Girdle Rotation

$VGRF_{PEAK}$ – Peak Vertical Ground Reaction Force

$VGRF_{TTP}$ – Time to Peak Vertical Ground Reaction Force

VGRI – Vertical Ground Reaction Impulse

CHAPTER I

INTRODUCTION

Acceleration and sprinting speed is of major importance to athletes competing in a wide variety of sports. The main objective for any running athlete is to cover ground quickly and this occurs by changing speed. Anytime there is a change in a speed there is acceleration. In sporting events when the athlete begins at rest and progresses into movement, the athlete is considered to be accelerating and this is often thought of a sprint.

Many authors have suggested that a track sprinting event is made up of different phases (Delecluse et al., 1995; Johnson & Buckley, 2001; Mero et al., 1992). Johnson & Buckley (2001) implied a 100 meter (m) sprint can be divided into three main phases of high acceleration, maximum velocity and steady state velocity maintained for the remaining distance. Though maximal velocity and maintenance phases have been researched more extensively than the acceleration phase, the acceleration phase is unique in that seventy-five percent of velocity is achieved during the first seven steps (Seagrave, 1996). In addition, the development of acceleration has been considered to have the greatest influence on the overall sprint result (Stein, 1999).

Sprint starts and the early acceleration phases (10 m) are a critical component to overall performance in track events (Coh et al., 1998; Harland & Steele 1997). It has been reported that the winners of 100 m sprint events would have also won if the race were only 10 m. Thus a majority of sprint training focuses on the acceleration phase (Ae et al., 1992; Ferro et al., 2001; Moravec et al., 1988; Muller & Hommel, 1997) and as a result some elite sprinters focus a majority of training on the acceleration phase (Ae et al., 1992; Moravec et al., 1988). While maximum velocity is a significant component to sport performance, the ability of the athlete to accelerate and develop greater acceleration during competition is also of importance (Douge, 1988; Murphy et al., 2003; Reilly, 1997; Hansen, 2006).

Many sports other than track involve sprinting. Soccer, for example, employs a sprint every four to five seconds during a match (Bangsbo et al., 1991) and the athletes may never reach maximal velocity during every speed burst. Consequently, the acceleration phase could be the most important component of shorter sprints occurring in team sports and is why it is the focus of this project. Once a sprint is initiated, the acceleration begins where the overall goal is to increase horizontal velocity. Horizontal velocity is a product of an athlete's step length and step frequency (Donati, 1996; Hay & Nohara, 1990; Hunter et al., 2004). Step length is the distance between successive foot strikes (right foot contact to left foot contact) while step frequency is the number of steps a person takes per unit time. Hunter et al. (2004) identified three key components within step length and frequency as stance and flight distance and time (Atwater, 1982; Lockie et al., 2003; Mero, 1988; Mero et al., 1983; Moravec et al., 1988; Murphy et al., 2003).

In addition to the kinematics of sprinting, identifying sprinting kinetics allows for greater understanding of overall mechanics such as utilizing ground reaction force (GRF) to predict acceleration (Hunter, 2005). GRF is an application of Newton's Third Law of

Motion, which states that for every action there is an equal and opposite reaction and as such when a sprinter exerts a force into the ground, the ground exerts an equal and opposite force to the foot (Hamil & Knutzen, 1995). The GRF is comprised of the vertical, medial/lateral, and anterior/posterior forces and when analyzed in relation to body mass, represents the acceleration of an athlete (Hunter et al., 2005).

A more successful sprinter must be able to exert greater GRF while limiting ground contact time (Alexander, 1989; Kunz and Kaufmann, 1981; Weyand et al., 2000) suggesting that both force and time play a significant role in acceleration. It is the product of force and time that represents a change in velocity, also known as impulse (Hunter et al., 2005). Braking and propulsive force serve important roles with relation to overall sprint performance and both Mero and Komi (1986) as well as Wood (1987) suggested that sprinters should inhibit braking force and enhance propulsive force. Limited research is available on the effects of GRF during the acceleration phase of a sprint.

The lumbopelvic-hip complex (LPHC) is the anatomical connection between the spine, pelvis and femur, which is highly coordinated in bipedal locomotion (Scache et al., 1999). The LPHC serves as a complex where 29 different muscles, which are responsible for stabilizing, transferring, reducing, and producing force during closed chain movements, attach (Goodman, 2007). However, while the LPHC is an important aspect in sprinting mechanics, literature is limited regarding its contribution.

The pelvic girdle, or pelvis, forms a connection between the spine and lower extremities (Vleeming & Stoeckart, 2007). The pelvis consists of right and left pelvic bones comprised of the sacrum, ilium, ischium and pubis joined by the pubic symphysis. The pelvis functions to support the weight of the body through the vertebral column, transmits ground reaction forces to the vertebral column during gait, and provides an attachment site for the musculature of the LPHC (Lippert, 2011). A primary muscle of the pelvic girdle is the gluteus maximus (GM). The GM originates on the crest of the ilium and inserts on the lateral surface of the greater trochanter of the femur (Floyd, 2009). The GM functions to extend, externally rotate and adduct the hip (Floyd, 2009). During gait, the GM is active in the role of hip extension as well as pelvic stabilization. GM activation begins eccentrically in late swing phase to negatively accelerate the thigh. Cavanagh (1990) suggested that it may assist in stabilizing the pelvis when extending the hip during the early stance phase.

The musculature of the LPHC provides the anatomical link between the upper and lower extremities. Examining the activation of the musculature about the LPHC in sprinting has been explored, but data are lacking when examining specific phases of sprinting. Mero & Peltola (1989) have reported that a 4.8% higher sEMG activity in the adductor magnus and GM occurs during the stance phase in acceleration when in comparison to the maximum velocity phase. The finding of Mero & Peltola (1989) has dramatic implications as the stance phase is where the maximal propulsive force for the athlete occurs, though Delecluse (1997) hypothesized that the difference in muscle activity between the acceleration and maximum velocity phase was due to increased trunk lean during the acceleration phase. With an increase in trunk lean during the acceleration phase, it is hypothesized that the sEMG activity of the GM will increase to provide stability to the pelvis during the stance phase in addition to providing greater propulsive forces with the center of mass well anterior of contact. However, the cause of the increased GM activity is unclear.

The latissimus dorsi (LD) is the largest muscle and serves as a bridge between the pelvis and the shoulder with attachment sites at: the posterior crest of the ilium, back of the sacrum, the spinous processes of the lumbar and lower six thoracic vertebrae (T6-T12) to the medial side of the intertubular groove of the humerus (Floyd, 2009). Based on the function of the LD, it is commonly only viewed as an internal rotator of the glenohumeral joint. However considering the whole geometry of the LD with its attachment on the pelvis, it is speculated that it could also play a role with the pelvis. For example, it has been suggested by David Weck, inventor of the Bosu ball and trainer for the world known sprinter Tyson Gay, that an arm motion called "spiraling" may be advantageous during sprinting. Spiraling involves pronation/supination at the proximal radio-ulnar joint, followed by internal rotation at the shoulder; therefore the role of the LD would seem to be paramount. The insertion of the LD on the spine and pelvis suggests that this muscle may have other responsibilities than just those at the shoulder (Patel et al., 2012; Vleeming & Stoeckart, 2007). Further support for the need for attention to the LD during locomotion, comes from an alternative approach to anatomy championed by Thomas Myers which considers not only the muscle but the fascial role as well (Myers, 2008). In regards to the LD, Meyers discusses the thoracolumbar functional line, which is the fascia connecting contralateral shoulders and hips. The thoracolumbar fascia is a collection of connective tissue found near the lower back (Myers, 2008).

Previous research has examined the relationship of the upper and lower extremity during gait, however only during walking (Elftman, 1939; Umberger, 2008). Data are lacking exploring the relationship between upper and lower extremity kinematics during sprinting. The LD is an extremely important muscle regarding arm swing, which is initiated at the shoulder. With the proximal attachment of the LD found on the humerus, the LD muscle activity during acceleration could indicate the arm swing function as it relates with bipedal locomotion. As muscles of the upper extremity have been mostly overlooked, this project will expand previous findings by including electromyography of the upper extremity, specifically the LD during bipedal locomotion acceleration.

The role of the upper extremity specifically the arms in walking, running and sprinting have received attention in literature (Bhowmick & Bhattacharyya, 1988; Elftman, 1939; Hay, 1993; Hinrichs, 1987; Umberger, 2008; Young, 2007) , however, the contribution of the arm swing to the acceleration phase of a sprint has been overlooked. There have been several theories as to the function of the arms during running yet, the role of the arm swing in sprinting remains a controversial topic. Arm movement has been considered to provide balance, stability, counter moments and even an excitatory effect on the lower extremity (Ferris et al., 2006; Korchemny, 1992). Young (2007) states that optimal arm swing is symmetrical and roughly matches the timing of the lower extremity. When running, the contralateral upper and lower extremities move synchronously, for example, as the right arm moves forward, the left leg moves forward.

Summary

There are no data available concerning the LPHC and upper extremity during sprinting, specifically the musculature of the GM and LD and their effects on upper extremity kinematics during sprinting. Therefore, the present study will analyze the kinematic, kinetic and muscle activity during the acceleration phase of a sprint with and without restricted arm and leg motion.

Statement of the Problem

The ability to run faster is one of the most sought after talents in sport. Not only is it important for track and field events, but most team sport athletes benefit from a greater acceleration in their respective sport positions. Literature investigating the acceleration phase of a sprint has neglected the role of the LPHC (Delecluse, 1997; Hunter, 2005; Mann et al., 1986; Mero & Peltola, 1989). The main goal of this study is to investigate the role of the LPHC and upper extremity during the acceleration phase of sprinting.

Statement of the Purpose

The purposes of this research are: 1) to determine the role of the LPHC during the initial acceleration of a sprint; and 2) to determine the overall effects of different arm and leg constraints by examining the spatial-temporal and joint kinematics, kinetics, and musculature activity, specifically the GM and LD, at different points during bipedal locomotion acceleration.

Hypotheses

Primary Objective – To determine the role of the lumbopelvic-hip complex, during bipedal locomotion acceleration under different arm and leg constraints.

1) Evaluate the muscle activity of the GM and LD during the acceleration of a 10 meter sprint using different arm and leg constrained conditions.

H_{01}: The constrained arms will cause a decrease in muscle activity of the LD muscle and an increase in muscle activity of the GM muscle.

H_{02}: The constrained legs will result in greater muscle activity of the LD muscle and a decrease in muscle activity of the GM muscle.

Secondary Objective – To determine the kinematics and kinetics during bipedal locomotion acceleration associated with different arm and leg constraints.

1) Evaluate the kinematic parameters of athletes performing an acceleration sprint with arm constraint conditions.

H_{01}: The arm constraint condition will result in shorter step length, shorter flight time, and longer stance time.

H_{02}: The arm constraint condition will result in decreased ankle plantarflexion, knee flexion, hip extension, and decreased posterior pelvic girdle tilt.

2) Evaluate the kinetics of athletes performing an acceleration sprint with arm constraint conditions.

H_{01}: The ground reaction force will decrease with lack of arm motion for participants.

H_{02}: The leg constraint condition will result in increased ankle plantarflexion, knee flexion and arm swing.

H_{03}: The propulsive impulse will decrease with lack of upper leg motion for participants.

Limitations

The limitations for the present study include the following:

1) Participants will complete all four conditions in one day.

2) No elite sprinters will be used as participants. Participants will be previous athletes who are currently recreationally active.

Delimitations

The delimitations for the present study include the following:

1) Participants will be required to wear compression clothing with retro-reflective markers placed bilaterally and surface sEMG electrodes unilaterally.
2) Participants will be between the ages of 19-30 years.
3) The participants will be assessed in the Sport Biomechanics Laboratory as rather than on a track or open field.

Definition of Terms

Acceleration: The rate at which velocity changes for an object.

Center of Gravity: The equilibrium point of the body where all three cardinal planes intersect.

Kinetics: Acting forces which produce a motion.

Kinematics: Measures which describe motion with respect to time.

Surface Electromyography: A non-invasive technique to measure electrical activity in muscles.

CHAPTER II

REVIEW OF LITERATURE

A vast majority of sporting endeavors involve bursts of speed. Those bursts of speed, in the running realm, are referred to as acceleration; however, a majority of running literature has focused primarily on constant velocity running. This project will attempt to further explain the acceleration phase of sprinting by investigating the influence of the lumbopelvic-hip complex during acceleration. Specifically, this study will consider the effect of constraining upper leg or arm motion during bipedal acceleration on the following: trunk and lower extremity kinematics; selected spatial-temporal measures; ground reaction force and impulse; and muscle activity of the gluteus maximus and latissimus dorsi muscles.

This chapter is divided into three sections: [1] examination of literature on the biomechanics of bipedal locomotion, specifically the kinematics and kinetics, [2] the role of the LPHC musculature, specifically the GM and LD, and [3] a summary of pertinent findings in kinematic, kinetic and muscle activation data of literature to the present study.

Biomechanics of a Sprint

A sprint is a short running event and includes a rapid increase in velocity. Every sprint, whether it is 10 meters or 100 meters, is comprised of an acceleration phase. When examining sprinting, the movement is often divided into phases of [1] acceleration, [2], maximum velocity, and [3] maintenance of velocity (Johnson & Buckley, 2001). The acceleration phase of any sprint begins with the motion and ends when maximal velocity has been reached. The acceleration phase is considered to have the greatest influence on the sprint results (Ae et al., 1992; Coh et al., 1998; Harland & Steele, 1997; Mero, 1988; Stein, 1999).

Locomotion velocity is the product of step length and step frequency while during running both are seen to increase as speed increases (Coh et al., 2006; Kunz & Kaufmann, 1981; Kyrolainen et al., 1999; Mann & Hagy, 1980; Mann & Herman, 1985; Mero et al., 1992). Athletes often manipulate step length, step frequency or both to enhance performance. Mann and Hagy (1980) researched the biomechanics of athletes walking, running and sprinting at constant velocities. The distinction between running and sprinting is typically marked by a greater speed while changing contact from hindfoot to the forefoot. In addition, sprinting incorporates longer step lengths when compared with running (Novacheck, 1998). Thirteen males; two sprinters, 5 joggers and 6 distance runners were used as subjects in a study conducted by Mann and Hagy (1980). It was revealed that step length, step frequency and horizontal velocity increased while contact time decreased when progressing from walking to running to sprinting (Mann & Hagy, 1980). In addition, Kunz and Kaufmann (1981) completed a study examining kinematic differences between 16 decathletes and 3 sprinters while performing in a 100 m race. Four strides (a stride is comprised of 2 steps) were captured at the 70 m point of the race. The authors chose the 70 m point for analysis because they felt that the athlete had completed the acceleration phase and were in the maximal velocity phase. Results revealed that the sprinters displayed longer strides and greater stride rates (stride frequency) when compared to the decathletes.

Kyrolainen et al. (1999) also observed kinematics at different running velocities of seventeen mid-distance runners. Along with most literature (Coh et al, 2006; Kunz & Kaufmann, 1981; Mann & Hagy, 1980; Mero et al., 1992), Kyrolainen and colleagues (1999) reported that as running speed increased, stride length and stride rate (frequency) increased. In an additional study, Coh et al. (2006) examined block starts and initiation of acceleration in a female international class sprinter. The sprinter performed 5 trials of a 20 m sprint while step length and frequency were recorded for the first 10 steps following the start. Step length progressively increased from step 1 to 10 (103 cm to 186 cm, respectively). Based on literature (Coh et al., 2006; Kyrolainen et al., 1999) the acceleration phase should yield progressive increases in step length.

A running gait cycle is often broken into contact time and flight time, which are determined by step length and rate (Hunter et al., 2004). Contact and flight times have been well versed in literature and have been found to decrease with increases in horizontal velocity (Atwater, 1982; Coh et al., 2006; Hunter et al., 2004; Kyrolainen et al., 1999; Lockie et al., 2003; Mann and Herman, 1985; Mero et al., 1992; Moravec et al., 1988; Murphy et al., 2003). Contact time is the period where either of the feet is in contact with the ground and has the following components: early, mid and late contact phases (Johnson & Buckley, 2001). Moravec et al. (1988) concluded that as horizontal sprint velocity increases, the proportion of contact time to flight time decreases.

Contact times at first and second ground contacts have been reported to be correlated with mean horizontal velocity over the initial acceleration phase of a sprint (r =

-0.65 & r = -0.44 respectively) (Atwater, 1982). Specifically, the contact times ranged from 170 – 230 ms for the first ground contact and 150 – 190 ms for the second ground contact after the start of the sprint. In addition, contact time was also reported by Hunter et al. (2004) at the 16 m mark of a 100 m sprint and was found to range from 124-125 ms. Furthermore, Coh and colleagues (2006), observed contact time to progressively decrease after a block start during the acceleration phase. The average for contact time at step 1 was 177 ms, while the contact time at step 10 was 110 ms, for a total decrease of 67 ms during the progression of the acceleration of the sprint.

Kyrolainen et al. (1999) also examined contact times of sprinters, but further divided this variable into braking and propulsive times. Braking time is the period in which the foot is exerting an anterior force on the ground, while propulsive time is the period over which the foot is exerting a posterior force on the ground. The study was completed on 17 middle-distance runners, 8 females and 9 males, who ran at constant speeds for 3 minutes on a 200 m indoor track. After the completion of 7 trials, a 15 minute recovery was given prior to 4 trials of 30 m sprints. Specific start techniques (whether block or standing starts) were not addressed. The researchers found that contact times (0.227 s to 0.115 s, $p<.001$) including braking and propulsive contact times decreased with increasing speed (.110 to .007 s, $p<.001$ & .117 to .09 s, $p<.001$, respectively). Along with the Kyrolainen et al. study, this study indicated that as running speed increased, contact times decreased (Kyrolainen et al., 1999).

As contact times have been shown to decrease with increases in speed, it is reasonable to conclude that flight times will increase and previous research has found this to be true (Atwater, 1982; Coh et al., 2006; Hunter et al., 2004; Mero, 1998; Moravec et al., 1988; Murphy et al., 2003). Flight phase is considered to be when the sprinter is no longer in contact with the ground and both feet are in swing phase. The flight phase can be broken down further into early, mid, and late flight phases (Johnson & Buckley, 2001). Early flight is when the body's center of gravity migrates upward and the leg moves posteriorly. Mid-flight is defined by the event of the leg under the body with knee flexed. Once the leg is in front of the body and the knee is extended to prepare for foot contact, it is considered late flight phase.

Several researchers have found that as horizontal velocity increases flight time increases (Atwater, 1982; Mero, 1988; Murphy et al., 2003; Moravec et al., 1988). In fact, during the acceleration phase, flight times at the first step ranged from 30 to 50 ms and 50 to 70 ms at second foot contact (Mero, 1988). Furthermore, Coh et al. (2006) discovered flight times to increase from step 1 to step 10 after a block start during the acceleration phase of a 20 m sprint. Average flight time at step 1 was 50 ms, while flight time at step 10 was 115 ms yielding an increase of flight time with an increase in velocity. A study by Hunter et al. (2004), considered flight time at a position 16 m from the beginning of the sprint and found that the flight time had increased to between 102-121 ms. These studies found that temporal components played a significant role during acceleration.

Literature has described rotation of the pelvis to be similar between running and walking (Mann, 1982; Mann, 1986; Scache et al., 1999), however the effects during accelerated locomotion have not been considered. Rotation of the pelvis about the vertical axis can be considered as transverse pelvic girdle rotation (Floyd, 2009). Right transverse pelvis rotation occurs when the right anterior superior iliac spine of the pelvis is rotated posteriorly, while left transverse pelvis rotation occurs on the contralateral side. However, interestingly, research involving the differences between walking and running in the transverse plane has also appeared to have opposing patterns (Novacheck, 1998; Ounpuu et al., 1990; Scache et al., 1999; Young, 2007) contradicting evidence for similarities between walking and running (Mann, 1982; Mann, 1986). The role of pelvic girdle rotation is not clear in biped locomotion and seems to serve different roles during walking, running and accelerating. For example, Young (2007) found that the pelvis rotates in the transverse plane during the contact phase of sprinting to increase stride length by nearly 5 cm (Young, 2007), however Scache et al. (1999) found that the pelvis does not serve to increases stride length in running.

The pelvis moves in all three planes during biped locomotion. Novacheck (1998) stated that the pelvis reaches maximal transverse rotation prior to foot strike and contralateral transverse rotation reaches peak levels during mid-contact phase. In addition, toward the end of contact, the pelvis begins ipsilateral transverse rotation to provide a neutral pelvis position near toe off. Ounpuu (1990) also found that children will achieve a neutral pelvis during late contact, prior to toe off. It is believed that this neutral pelvis position is advantageous for 2 reasons: a) neutral pelvis position infers that the thigh is in a neutral position, which is beneficial for efficient transfer of muscular forces in the LPHC and b) the neutral pelvis can assist with the contralateral limb preparation for foot contact. In addition to the role of the neutral pelvis, it is thought that contralateral transverse rotation of the pelvis may provide a decrease in braking ground reaction force, due to a quicker flexion moment at the contralateral hip, which would defer the possibility of a decrease in speed during the contact phase (Ounpuu, 1990). In light of these findings further research is needed to understand the role of pelvic girdle rotation and position during bipedal locomotion.

Previous research indicates that individuals of different levels of sprinting ability display significantly different joint angles during sprinting (Kunz & Kauffmann, 1981; Kyrolainen et al., 1999; Mann & Hagy, 1980; Mann & Herman, 1985; Mann et al., 1986; Murphy et al., 2004; Novacheck, 1998). In a comparison between faster and slower athletes, Murphy et al. (2003) found an 8% decrease in knee angle at takeoff of the third step for the faster athlete. The reduced range of motion of an athlete moving at a higher velocity may allow the lower limbs to accelerate more quickly which could increase sprint performance resulting from reduced knee extension or early activation of the hip flexors (Murphy et al., 2003). Kunz and Kaufmann (1981) noted smaller thigh angles at foot contact, greater thigh accelerations, and larger trunk inclinations when comparing world-class sprinters to decathletes in a 100 m bout.

Mann and Hagy (1980) investigated the biomechanical differences between athletes walking, running and sprinting and found increases in peak knee and hip flexion associated with greater speeds between walking, running and sprinting. While considering the ankle during running, peak dorsiflexion occurred during initial ground contact followed by increasing plantar flexion. However, during sprinting, initial contact with the ground was made by the forefoot and peak dorsiflexion occurred during the contact phase; plantar flexion followed this dorsiflexion with no apparent heel contact. Mann and colleagues (1980) also reported that as the speed of gait increases, the velocity about the hip and knee increased and that within 250 ms during sprint initiation the hip joint and knee joint changed 80° (320 degrees/second) and 65-70° (260-280 degrees/second), respectively. In addition, Kyrolainen et al. (1999) concluded that angular displacement of the ankle and knee joints decreased during ground contact when overall speed was increased with push-off yielding an increase in peak angular velocities.

To the author's knowledge, very few studies regarding the trunk angle within bipedal acceleration have been completed (Atwater, 1982; Kunz & Kaufmann, 1981), however research involving walking and running at constant velocities have been researched more extensively (Thorstensson et al., 1984; Thurston et al., 1981; Scache et al., 2002; Wank et al., 1998). Specifically, during leg extension, the lower spine extends while the upper spine rotates anteriorly with the arm to maintain balance during locomotion (Scache et al., 1999). Rotation of the trunk in the sagittal plane about the medial/lateral axis is known as flexion/extension. Specifically, it has been determined that at the 70 m mark of a 100 m dash, world class sprinters exhibit a larger trunk angle, which contributes to greater trunk extension, in comparison with decathletes (Kunz & Kauffmann, 1981). Trunk angle is defined as the midpoint along a line between the hip joints and C7 prominens. Wank et al. (1998) completed a study investigating 10 recreationally trained male individuals and examined the kinematic differences between participants running at 4 and 6 m/s over ground and on a treadmill. Data indicated that the runners achieved average amplitude of 10.2° and 8° of trunk flexion while running at 4 and 6 m/s, respectively. As a result, it was concluded that higher speeds yield less trunk flexion (Wank et al., 1998).

Furthermore, Thorstensson and colleagues (1984) investigated trunk motion in 10 males during walking and running on a treadmill, while participants completed the conditions of varying increment speeds set by the experimenter on a motorized treadmill.

Speeds indicated for walking and running ranged between 1.0-2.5 m/s and 2.0-6.0 m/s, respectively. Trunk flexion progressively increased from 6° to 13° with increases in incremented speeds. During walking, peak trunk flexion occurred during the beginning of support phase, while during the running trials; peak extension occurred at this point (Thorstensson et al., 1984). It was hypothesized that changes in trunk motion at higher speeds are due to mechanical conditions such as changes in center of gravity (CoG). Specifically, during locomotion it is advantageous to have the center of gravity in front of the propulsive limb. During walking this occurs during a time of dual support while the trailing limb is pushing off and the leading limb is making contact. In running however, foot contact occurs following a flight phase and the swinging forward of the soon-to-be lead leg would require an off-setting force of trunk extension (Newton’s third law) (Hinrichs, 2005).

While viewing a sprinters start in the sagittal plane, the trunk has a great amount of forward lean, or flexion, after leaving the blocks. After this initiation, the trunk progressively becomes more upright as measured from the vertical as the sprint continues. Atwater (1982) further examined trunk lean (as measured from the horizontal) after the start at the first, second, third and fourth contact phases in 8 national level sprinters and found the angles to be 24°, 30°, 37°, and 47° respectively. During the acceleration phase the trunk flexion angle decreases bringing about a more upright posture as the center of gravity progressively moves posterior as speed increases. The position of the CoG with regard to foot placement changes during the first several steps. At movement initiation, the CoG is in front of the contact point of the foot with the ground. Towards contact phase of the third step, the CoG is behind the point of contact (Mero et al., 1983).

Literature involving axial trunk rotation is scarce (Thurston et al., 1981; Scache et al., 2002). Thurston et al. (1981) examined 22 male participants walking at a comfortable speed and found 8.3 and 11.2 degrees of axial and transverse pelvis rotation. In addition, Schache et al. (2002) examined angular kinematics of 20 males running at 4 m/s on a treadmill. Results revealed a low correlation between axial rotation of the trunk and pelvis (r=.37), and high correlations between trunk flexion/extension and anterior/posterior pelvis rotation (r=0.75) (Schache et al., 2002). Schache et al. (2002) indicated that the low correlation between the pelvis and the trunk in the transverse plane suggests that the pelvis and trunk serve different functions and is a basic premise of the present research.

When an athlete is sprinting, ground reaction forces are acting on the athlete (Hunter, 2010) and can dictate acceleration (Hunter, 2005). To increase horizontal velocity, an athlete must have the ability to increase the amount of force applied to the ground and decrease the amount of time over which this force is exerted. Vertical and horizontal (forward) ground reaction forces have received the most attention in literature (Hunter, 2005; Hunter, 2010) on sprinting yet only a few studies have investigated ground reaction force during acceleration.

Horizontal GRF can be divided into braking and propulsive forces. Braking forces occur in the opposite direction of the movement while propulsive forces are directed with the forward motion (Young, 2007). Athletes should attempt to minimize braking GRF and maximize propulsive GRF in attempt to enhance forward motion. Mero (1988) conducted a study on 8 male sprinters that investigated the force-time characteristics during acceleration and found that braking forces occur in excess when initial contact occurs anterior to the center of mass. Excess braking forces cause a decrease is horizontal velocity occurs which inhibits performance. There is limited research examining the application of braking forces to sprint running but Merni et al. (1992) indicated that the fastest participant studied had lower braking GRF when compared to the other participants in the study. The study included 3 elite athletes performing 3 trials of a 60 m and 100 m dash sprint from a starting block. Peak braking forces of 1.2 x BW in the vertical direction and -0.7 x BW in the horizontal direction were reported for an advanced sprinter (Merni et al., 1992) demonstrating the significance of lower horizontal braking forces in acceleration performance. Merni et al. (1992) also considered the duration of the propulsive phase and found that there were significant differences between the 3 elite subjects, finding that the propulsive phase was inversely related ($r = -.96$) to overall sprint performance (Merni et al., 1992), indicating that greater propulsive phases yielded lower sprint times. Mero (1988) also found a significant correlation between propulsive GRF and horizontal sprint velocity during the beginning of the acceleration phase ($r = 0.62 – 0.69$), indicating that larger propulsive GRF yielded greater velocities.

The vertical GRF represents the ability of the athlete to slow the vertical velocity downward of the center of mass at initial contact and then reverse it at toe off (Hunter et al., 2010). Vertical GRF has been researched extensively (Bohn et al., 1998; Hunter et al., 2010; Keller et al., 1996; Mero, 1988; Weyand et al., 2000) with the general findings indicating that individuals running at high velocities produce large vertical GRF. Bohn et al. (1998) conducted a study that investigated the kinetics of maximal velocity sprinting in 24 runners, (21 male; 3 female) sprinting at maximum velocity. Results indicated that more advanced sprinters produced greater horizontal and vertical GRF at maximal velocity, 1.1 x BW and 3.6 x BW, respectively, when compared to a less advanced sprinters (0.6 x BW and 3.3 x BW respectively). Researchers concluded that a more powerful acceleration is evident in the advanced sprinter, who had greater vertical and horizontal GRF. In addition, Weyand et al. (2000) suggested that during maximal velocity of sprinting, vertical GRF plays a crucial role in evaluating overall sprint performance. Researchers examined 33 participants with varying sprinting abilities while running at top speeds on a treadmill. Data indicated that vertical GRF was greater with the highest sprint velocities. The authors concluded that a faster sprinter achieves maximal velocity by decreasing contact time and increasing vertical force (Weyand et al., 2000). Furthermore, Keller et al. (1996) examined 13 male and 10 female participants walking and running. Velocities for walking were increased incrementally between 1.5 and 3 m/s and for running between 3.5 and 6 m/s. Results indicated that vertical GRF increased linearly during walking and running from 1.2 BW to 2.5 BW.

Mero (1988) claimed that the greatest vertical GRF forces are developed during the acceleration phase, once the athlete has reached near maximal velocity. Vertical GRF of the early acceleration phase, represented by the first 3 m, were 1.6 BW, while vertical GRF after the 35 m mark of the sprint were 2.3 BW. Though not stated by the researchers it is reasonable to conclude that during the acceleration phase, the athlete will attempt to generate greater horizontal GRF than vertical. This is further supported by the relatively small flight times noted during initial accelerations.

Ground reaction impulses (GRI), are the product of the GRF and the duration over which it is exerted occurring in all directions, primarily focused in the horizontal and vertical directions. It has been reported that horizontal GRI significantly correlates with horizontal sprint velocity after a block start thus allowing one to conclude that as horizontal GRI increased, horizontal velocity increased as well (Mero, 1988).

Hunter et al. (2005) examined the performance of 36 participants in a maximal effort sprint. The main focus of the study was to test the GRF and GRI during acceleration at 16 meters. Each athlete performed 7 to 8 sprints with a result of 4 or 5 successful trials, which were determined by the participants' foot completely contacting the force plate. Vertical, horizontal, braking and propulsive impulse data from the force-plate were calculated. Results revealed a strong trend for faster athletes to produce greater propulsive impulses. Hunter and colleagues (2005) reported a positive relationship between propulsive GRF and overall sprint velocity (r=0.65) but implied there was no support for a relationship between sprint velocity and braking impulse, which is surprising since Merni et al. (1992) indicated that the fastest sprinters demonstrated lower braking forces and greater propulsive forces.

Kinetics play an important role in human locomotion. How the body pushes against the ground is directly related to how the ground pushes against the body. While kinetic analysis of acceleration associated with running initiation is limited, extrapolations from previous work on running can be made. It is hypothesized that initially the braking impulse will be minimal as the horizontal velocity of the foot will be minimal, however the propulsive impulse will be large as the foot will be in contact with the ground longer and maximum force production will be desired. Therefore, the initial acceleration phase will provide this project with a unique window into the propulsive component of bipedal acceleration. For example, during acceleration it is anticipated that the runner will attempt to increase posterior/propulsive forces, minimize anterior/braking forces and demonstrate diminished vertical forces. Furthermore, the accelerating runner will produce large propulsive impulses (as there will be an attempt to maximize horizontal forces and the contact time will be extended).

The Lumbopelvic-Hip Complex

The lumbopelvic-hip complex (LPHC) is an anatomical connection between the spine, pelvis and femur (Scache et al., 1999; Vleeming & Stockart, 2007). With 29 muscles attaching to the LPHC, the function of this complex is very important is static posture and mobility. The CoG of the body is located within this complex, with movement initiating from the LPHC (Scache et al., 1999). Interestingly, the LD is the only muscle crossing the shoulder joint, which also attaches to the LPHC. Furthermore, the GM muscle is found on the posterior aspect of the pelvis, with proximal insertions on the crest of the ilium and sacrum while distally inserting on the lateral surface of the femur. The GM is responsible for extension and external rotation of the hip, and posterior pelvis rotation (Floyd, 2009). Mero and Peltola (1989) found that sEMG activity patterns were similar in the maximal velocity and maintenance phases of a 100m race but peak activity occurred during the acceleration phase. These findings suggest that while sEMG activity is similar between the two phases, there is no change in neural recruitment once the acceleration phase is complete and maximum velocity is achieved.

In addition to sEMG, indwelling EMG has also been utilized in research involving bipedal locomotion. A study completed by Montgomery et al. (1994) examined knee and hip muscles of 30 recreational runners running at 3 different speeds. The authors concluded that the rectus femoris muscle activated to assist hip flexion during the swing phase and the hamstrings and GM activated for eccentric action during hip flexion. In addition, Novacheck (1998) examined sEMG during running and found that a majority of the lower extremity musculature was active during the late swing and beginning of contact phases. Similar research conducted by Jonhagen et al. (1996) examined 9 sprinters during maximal velocity sprinting and it was determined that the GM showed peak activation just prior to and during foot strike. The authors concluded that the GM works to accelerate the thigh in the negative direction via eccentric muscle activity during the final part of swing phase (Jonhagen et al., 1996), thus slowing the thigh in preparation for foot contact.

Weiman and Tedo (1995) examined hip and knee musculature of 12 national champion sprinters at the 50 m mark of a 100 m sprint while running at maximal velocity. Results indicated hamstring activity to be more prevalent during late swing, support and early swing phases while a significant decrease in activation occurred during the contact phase. In addition, sEMG of the vastus medialis demonstrated a majority of activity during late swing and contact phases (Weiman & Tedo, 1995). The authors concluded that the GM supplied the necessary energy for forward propulsion during eccentric hip flexion and concentric hip extension. Furthermore, Kyrolainen, et al. (1999) emphasized the importance of the hip extensors in a study examining changes in muscle activity with higher running speeds. The study was conducted on 17 middle-distance runners on a 200 m indoor track running at constant speeds. GM and biceps femoris increased activation during the braking phase. Literature suggested that the hip extensor musculature is very important and muscle activity may increase as horizontal velocity increases as well by limiting braking GRF and GRI during contact (Kyrolainen et al., 1999). Kyrolainen and colleagues (1999) examined 2 male sprinters running a 100 m bout. It was concluded that during foot contact there was a 4.8% increase in sEMG activity in the GM, vastus lateralis, biceps femoris, gastrocnemius and tibialis anterior during the acceleration phase when compared to the maximal velocity phase (Kyrolainen et al., 1999).

Previous research examined muscle activity of the GM during walking, running and sprinting (Jonhagen et al., 1996; Kyrolainen et al., 1999; Mero & Peltola, 1989; Montgomery et al., 1994; Novacheck, 1988; Weiman & Tedo, 1995) , however few have focused on the acceleration phase (Coh et al., 2009). Coh et al. (2009) evaluated sEMG of the lower extremity musculature at the first 2 steps after the block start while the sprinter was accelerating. Results indicated peak GM activity during the swing phase (Coh et al., 2009). While the results are interesting, sEMG was only taken immediately following the block start, more research involving the following steps is crucial to understanding the full muscle activation sequence and importance of the GM during acceleration.

The LD is a large muscle on the back with multiple attachment sites including the posterior crest of the ilium, back of the sacrum, and the spinous processes of the lumbar and lower six thoracic vertebrae (T6-T12) and the medial lip of the intertubercular groove of the humerus (Floyd, 2009). In addition, the LD and external obliques are linked together via insertions found at the ribs (Kendall et al., 2005). The roles of the LD as a shoulder extensor, adductor and internal rotator have been well documented (Floyd, 2009; Baechle & Earle, 2000; Lehman et al., 2004). However, previous research focusing on the LD has considered the muscle to have a role during pelvic girdle and trunk motions and possibly affect the techniques of the upper extremity in athletic performance (Kendall, 2005; Patel et al., 2012; Vleeming & Stoeckart, 2007; Weck, 2011).

Few have studied muscle activation of the LD while running (Cappellini et al., 2006; Ivanenko, 2004). For example, Cappellini et al. (2006) examined motor patterns of the LD during walking and running in 8 subjects, 6 males and 2 females, on a treadmill. Cappellini and colleagues recently examined upper and lower extremity muscle activations during treadmill walking and running. It was reported that the LD had greater activation as the speeds increased. In addition, Ivanenko (2004) examined muscle activation during treadmill running and also reported increased LD at higher speeds. Similar conclusions were presented by Jones et al. (2009) who investigated sprinters, and advised that shoulder muscles should be relaxed once an upright position is adopted as the runner achieves maximum velocity.

Patel et al. (2012) revealed that the LD contributes to pelvic girdle and trunk motions. The researchers examined sEMG during isometric shoulder internal rotations at different pelvis and trunk positions. It was concluded that the LD contributes greatly to pelvic girdle and trunk rotations. An upright posture and posteriorly rotated pelvic girdle promotes greater front side mechanics and limits backside mechanics, which are important to the efficiency of the sprint (Young, 2007). Posterior pelvis rotation implies that the LD has been lengthened due to the distal attachment on the pelvis moving inferior, therefore implying the stretch-shortening cycle providing elastic potential energy to be released when the muscle shortens during arm swing. This suggests that the LD may have a larger role in bipedal locomotion than previously noted in literature (Patel et al., 2012).

In addition to LD muscle activation, muscular strength has also been examined (Patel et al., 2012). Patel and colleagues concluded that altering pelvic girdle position would alter glenohumeral internal rotation force development. Seventeen current resistance training males performed isometric glenohumeral internal rotation at positions of 10 degrees contralateral pelvic rotation (CPR) and ipsilateral pelvic rotation (IPR); 20 degrees CPR and IPR; full CPR and IPR. A significant difference between the force in the neutral condition and the 10 and 20 degrees of IPR was found. It was suggested that altering pelvic girdle position does alter force development, which could have implications for the role of the LD during bipedal acceleration.

The LD is considered as one of the largest muscles in the body, not only because of its natural anatomy but its synergistic interaction with surrounding musculature via connective tissue (Myers, 2008; Vleeming & Stoeckart, 2007). The thoracolumbar fascia (TLF) (Figure 1) aids the LD and GM in providing a pathway for energy transmission between the LPHC (Vleeming & Stoeckart, 2007). Vleeming & Stoeckart (2007) state that the LD and GM exert a contralateral effect with other musculature on the vertebral column and pelvis by developing tension during motion with assistance from the counter-rotation of the trunk and flexion of the arm. This contralateral effect may be beneficial for an athlete accelerating due to the motions occurring at the trunk and pelvis. In addition, trunk rotation has been shown to activate the LD (Kumar et al., 1996; Vleeming &

Stoeckart, 2007; Patel et al., 2012), and thus possibly affecting athletic performance during acceleration. Contrary to the findings of Vleeming & Stoeckart (2007), Kumar et al. (1996) and Patel et al. (2012), Bogduck (1998) claims that the LD is designed to move the arm and the ability of the LD to stabilize the sacro-iliac joint in the pelvis with TLF is not justifiable.

Fascia is a layer of connective tissue, which surrounds the musculature throughout the body. The novel sprinting technique coined "spiraling" (Weck, 2011) is based upon the role of fascia in bipedal locomotion and suggest that the pronation at the proximal radio-ulnar joint followed by internal rotation at the shoulder as the arm is driven back improves the sprint velocity of an athlete by utilizing the fascial link (Weck, 2011). The fascia line most involved with sprinting is the thoracolumbar functional fascia line (TFL). The TFL line runs from the shoulders, to the contralateral hips, and lower back. Traditional fascia serves to increase energy transfer between connecting musculature (Myers, 2008; Bogduck et al., 1998; Mooney et al., 2001; Vleeming & Stoeckart, 2007; Dorman, 1992). The TLF plays an integrating role in trunk rotation and stability of the lumbar vertebrae and pelvis and can be considered as a prime element for athletic ability. The LD and contralateral GM muscles are coupled via the TLF and can provide tension to the surrounding connective tissue (Vleeming & Stoeckart, 2007). Vleeming & Stoeckart (2007) concluded that the posterior layer of the TLF could play an important role in force transmission between the legs, pelvis and spine, in addition to trunk stabilization and rotation. Furthermore, the TLF develops tension during motion with assistance from the counter-rotation of the trunk and glenohumeral flexion with direct implication for human gait (Vleeming & Stoeckart, 2007).

Accordingly, Dorman (1992) compared this mechanism via extending a finger and releasing it supposing that the energy is stored by the fascia surrounding the musculature. For example, this mechanism can be completed by pulling the finger towards the dorsal surface of the hand, thus decreasing the angle between the finger and hand prior to releasing it. This research involving a stretch of the tissue followed by shortening indicates the stretch-shortening cycle can be advantageous with certain movements. Moreover, the mechanism can be related to the motions occurring between the pelvis and trunk and pelvis and hips, where pelvis stabilization occurs followed by the releasing of the complex during locomotion.

Mooney et al. (2001) have studied the relationship between the TFL, LD and contralateral GM during axial rotation exercises and walking (Mooney et al., 2001). Fifteen healthy individuals, 12 right handed, walked on a treadmill at a normal pace. It was reported that the right GM muscle had lower activity in comparison with the left. The authors concluded that the functional relationship between the LD and GM could be confirmed due to the abnormal hyperactivity of the GM and greater activity of the contralateral LD in the participants (Mooney et al., 2001). It is interpreted as the opposing relationship of the two muscles correlates with shoulder and pelvic girdle rotation in gait due to the attachment sites of the LD and GM found at the humerus and pelvis and pelvis and femur, respectively. In addition, results indicated that the right LD was significantly more active than the left LD during right trunk rotation; however, the right GM was less active than the GM on the left side during right trunk rotation (Mooney et al., 2001) implying the activation of the muscles due to trunk rotation during locomotion, specifically increase in trunk flexion during acceleration.

Due to the coupling between the LD and contralateral GM, categorizing specific roles of the LD with limitation to the shoulder diminishes consideration of the LD function with the pelvis and vertebral column (Patel et al., 2012; Vleeming & Stoeckart, 2007). It is known that the GM is active during eccentric hip flexion to slow the leg prior to initial contact and concentric extension of the hip during contact (Cavagna, 1990; Jonhagen et al., 1996; Kyrolainen et al., 1999; Novacheck, 1998). In addition, it is understood that the LD is active during shoulder extension, which drives the arms posteriorly (Baechle & Earle, 2000; Floyd, 2009; Lehman et al., 2004). However it is not known as to how the GM and LD work independently at the pelvis and in conjunction through the TLF link.

Figure 1. The thoracolumbar functional fascia line as it connects the upper and lower extremities (Myers, 2008).

Sprinting involves a coordination of movements between the upper and lower extremities. However, research involving both sprinting and arm motion has been scarcely examined. Though it seems obvious that the lower extremity performs a majority of the work necessary for locomotion, the function of the arms are not clear. Literature reveals contradicting theories regarding the role of arm swing while walking and running with limited research on sprinting. Elftman (1939) proposed the first hypothesis regarding arm swing in walking suggesting that the function of the arm movement was to counteract the angular momentum, created by the legs, about the vertical axis with the shoulder muscles developing this counter-action. Studies regarding the metabolic cost of walking suggest that an increase in metabolic cost occurs when arm swing is inhibited (Umberger, 2008; Collins et al., 2009). In addition, Collins et al. (2009) found that an increase in metabolic cost occurs during anti-normal arm swing, where angular momentum is offset. Authors stated anti-normal arm swing is where the arm swings forward with ipsilateral hip. Furthermore, the shoulder muscles have been stated to actively drive the arms during gait (Ballesteros et al., 1965) while more recent studies suggest the arm swing occurs passively from shoulder musculature (Collins et al., 2009; Pontzer et al., 2009). A main argument involves the question whether the arm swing is active or passive. Passive arm swing is a result of trunk rotation, while active arm swing requires musculature activation from shoulder muscles to produce the desired arm swing (Ballesteros et al., 1965; Elftman, 1939). It has been proposed that the arms act as passive dampers to limit rotation of the trunk during walking and running (Pontzer et al., 2009). Pontzer et al. (2009) examined treadmill walking and running during weighted and unweighted arm swing conditions. The researchers concluded that the arm swing is a passive response as the arms decreased trunk torsion rotation, and were further supported by the action of the deltoids which stabilized the shoulder in accordance with passive arms swing hypothesis (Ponzter et al., 2009). The anterior deltoid fired when the arms moved posteriorly, while the posterior deltoid fired when the arms moved anteriorly, suggesting most deltoid contractions were eccentric. However, only deltoid musculature were examined.

Mann et al. (1984) investigated the angular kinematics at the shoulder during in elite male and female sprinters while competing in 100 m to 400 m events during 1982-1983. It was determined that forearm angles change throughout the running gait cycle. Specifically, minimum angles were seen at full front and full back position, while maximum angles were found during each stride at the midpoint of the arm swing (Mann et al., 1984). When investigating the angles between the arms and trunk, it was reported that the arm makes a 45° angle with the trunk when anterior and an 80° angle with the trunk when posterior during each stride with excessive arm action signaling over-striding (Mann et al., 1984). While the debate regarding arm swing at the shoulder has yielded opposing views the role of flexion and extension at the elbow is less clear and may simply be a passive response to the mechanical loading of the whole body.

While considering the role of arm swing to running gait, the novel sprinting technique coined “spiraling” (Weck, 2011) should be discussed. Spiraling involves a rotary component to the usual arm swing motion. In particular, the arm supinates at the proximal radio-ulnar joint when the arm is forward which is followed by internal rotation at the shoulder as the arm is driven back. During spiraling, pronation that is occurring when the humerus is extended aids ipsilateral hip flexion. This pronation will lead to internal rotation of the shoulder and the “recoil” of shoulder musculature as the musculature prepare to assist in humeral flexion, which Weck (2011) believed will release the tension stored in the fascia. Support for this notion comes from previous work by Novacheck (1998) who suggested that externally rotating the shoulder at the top of the arm swing stretches the LD and helps create potential energy in addition to aiding correct posterior pelvic girdle rotation, which is vital for sprint performance.

Mann (1981) examined 15 males during maximal velocity sprinting and reported that the moments at the upper extremity were relatively small and based on this finding concluded that the role of the arms in the sprint is only for balance. In addition, Hinrichs (1987) found that the arms may contribute only up to 10% of the total vertical propulsive forces and impulses when considering ground reaction forces. Hinrichs (1987) examined 10 male recreational runners while running on a treadmill and the arms were found to generate an alternating pattern. Further, it was suggested that the arm action is important in the conservation of angular momentum while providing the trunk axial rotation necessary to counteract leg motion, due to the inverse relationship between the upper and lower extremities (Hinrichs, 1987).

Various team sports require accelerations throughout competition with different arm constraints. Grant et al. (2003) investigated the effects of different ball carrying methods of rugby players on sprint speed. Forty-eight rugby players completed trials of 20 m sprints with different ball holding conditions. Results indicated that conditions where the athlete had to run with the ball in both hands yielded a decrease in sprint velocity when compared to those conditions where the arms were free to move (2.62 s and 2.578 s, respectively). Furthermore, Sayers (1998) found that when a rugby player carried the ball in one hand, there is a reduction in stride length and an asymmetrical pelvic rotation. With both hands of the athlete on the ball, the arms were not allowed to move in opposition to the leg swing and because of that, the athlete may not be able to take advantage or utilize the energy storing potential of the fascia, specifically the TLF involving the LD and GM.

Widowski and Gittoes (2012) investigated sprint techniques of field hockey players with and without carrying a field hockey stick with both hands. Eighteen currently competing male field hockey players completed six sprint trials, constrained and unconstrained. Results indicated that stride length and stride frequency were maintained in both conditions because the stick was longer and the athletes could get enough forward motion with each side of the stick. However sprinting without constraints had a greater velocity when compared to constraints (.10 m/s), suggesting athletes had to alter running form and velocity when their arms were constrained. In addition, Frere et al. (2009) examined the influence of pole carrying on the kinematics of a pole vaulter sprinting to the vault position. Eight currently training athletes performed six total trials (three with pole and three without) of 18 m sprints. Results indicated that velocities decreased when the athlete was running with the pole when compared to running without the pole (6.17 m/s and 6.57 m/s, respectively). In addition, hip and knee flexion decreased through the sprint with arm constraints yielding a decrease in stride length.

During walking, the arms swing in opposition to the legs (Elftman, 1939). Evaluating the overall impact of arm swing suppression during locomotion is a difficult task. Umberger et al. (2008) studied the effects of arm swing suppression with 8 participants, 5 male and 3 female, while treadmill and ground walking. Researchers constricted arm swing by having the participants fold their arms across the chest such that each hand was supported on the contralateral elbow. Results indicated energy expenditure was increased with arm swing restriction. In addition, kinematic variables regarding joint angles, angular velocities and ground reaction forces were “nearly identical” between conditions of walking with and without arm motion (Umberger et al., 2008). It was concluded walking models without articulated arms may be justifiable in model based research.

In addition to arm swing restriction on walking, the effects on running should be considered to provide beneficial knowledge on the overall impact of this condition. Pontzer et al. (2008) examined the function of arm swing while treadmill walking and running. The study was completed with three conditions, a normal control, arms folded across the chest of the participant, and arm weights at the elbow. Interestingly, anterior and posterior deltoid activity occurred simultaneously (Pontzer et al., 2008), in opposition with previous claims made by Ballesteros et al. (1965), which found an alternating sEMG activity of these muscles. It was concluded that arm swing is derived from trunk rotation. Consequently, it was also concluded that upper extremity motion is generated by the lower extremity and the arms acted as passive dampers to reduce trunk rotation (Pontzer et al., 2008). However the maximum speed for the participants running was limited to 3 m/s and all trials were completed on a treadmill providing little benefit to athletes competing in over ground sports or during changing speeds.

In addition, Miller et al. (2009) examined lower extremity kinematics and kinetics with different conditions of upper extremity restrictions (a control condition with normal arm swing, arms crossed anteriorly to the chest, and arms crossed behind the back) while running at 5K training pace. The results demonstrated that peak vertical GRF decreased and duration of contact phase increased with the suppression of arm swing. In addition, peak hip flexion increased by 1.7°, peak knee flexion increased by 4° when the arms were restricted, however timing across joint actions were not significantly affected by arm restriction. It was concluded that arm swing suppression affects GRF and lower extremity joint kinematics (Miller et al., 2009).

Summary

Research suggests that the acceleration phase of a sprint has the greatest influence on the overall sprint results (Stein, 1999; Ae et al., 1992; Coh et al., 1998; Harland & Steele, 1997; Mero, 1988). In attempt to achieve maximal acceleration, many biomechanical aspects are altered. It has been reported that as contact time decreases, step length and flight time increase during higher velocity locomotion and acceleration (Atwater, 1982; Coh et al., 2006; Hunter et al., 2004; Lockie et al., 2003; Kunz & Kaufmann, 1981; Kyrolainen et al., 1999; Mann & Hagy, 1980; Mann & Herman, 1985; Mero, 1988; Mero et al., 1992; Moravec et al., 1988; Murphy et al., 2003). In addition, research has indicated a decrease in braking contact times while propulsive contact times increase during acceleration (Kyrolainen et al., 1999).

Joint kinematics are an important variable in sprinting mechanics. For example, the role of the pelvis in bipedal locomotion is not clear and is under debate due to the contradicting evidence found in literature regarding walking and running at constant velocities (Mann, 1992; Mann, 1986; Novacheck, 1998; Ounpuu et al., 1990; Scahce et al., 1999; Young, 2007); however results corresponding to acceleration are limited and need further investigation. Increases in peak knee and hip flexion and ankle plantarflexion have been observed as testing speeds have increased as well (Kunz & Kauffmann, 1981; Kyrolainen et al., 1999; Mann & Hagy, 1980; Mann & Herman, 1985;

Mann et al., 1986; Murphy et al., 2004; Novacheck, 1998). In addition to lower extremity kinematics, trunk position has also received attention. Atwater (1982) determined that trunk lean progressively increases after the start during acceleration. This is due to the position of the CoG of the athlete changing during the first several steps of acceleration. The CoG moves behind the point of contact as speed increases during acceleration followed by a progressive increase in trunk angle (Atwater, 1982).

In addition to limited research on kinematics during acceleration, research involving the kinetics during this increase in speed is limited as well (Hunter, 2005; Merni et al., 1992; Mero, 1988; Young, 2007). It can be concluded that minimizing braking GRF and maximizing propulsive GRF enhances acceleration and forward motion (Merni et al., 1992); however contradictory evidence from Hunter et al. (2005) suggests that there is no relationship between sprint velocity and braking impulse. Moreover, it has been determined that a more powerful acceleration is associated with greater vertical and horizontal GRF.

The LPHC is the anatomical connection between the spine, pelvis and femur and is highly coordinated in stability and mobility (Scache et al., 1999; Vleeming & Stockart, 2007). sEMG research involving the GM and constant velocity locomotion is evident in literature (Jonhagen et al., 1996; Kyrolainen et al., 1999; Mero & Peltola, 1989; Montgomery et al., 1994; Novacheck, 1998; Weimann & Tedo, 1995); however research focused on the acceleration phase is limited (Coh et al., 2009). It is well understood that sEMG activity is greater with increases in speed, however literature is limited directly focusing on acceleration, specifically with arm and leg constraints. In addition, LD sEMG activity has been investigated during walking and running (Cappellini et al., 2006;

Ivanenko, 2004) but the effects with acceleration are scarce and require further investigation. It is evident that the GM and LD function at the pelvis through the TLF (Myers, 2008; Mooney et al., 2001; Vleeming & Stockart, 2007; Patel et al., 2012; Patel et al., 2013) which serves to assist the LPHC during locomotion. However, the effects on acceleration are limited.

The function of arm swing during locomotion is a debated topic in literature with some proposing the passive swing model (Collins et al., 2009; Pontzer et al., 2009) and others proposing the active swing model (Ballesteros et al., 1965; Cappellini et al., 2006; Elftman, 1939; Novacheck, 1998). To aid this debate several studies have considered locomotion with the arms constricted (Frere et al., 2009; Grant et al., 2004; Miller et al., 2009; Umberger et al., 2008; Widowski & Gittoes, 2012). It can be concluded that constricting arm motion decreases step length and horizontal velocity, with increases in peak knee and hip flexion. The adaptations occurring in the body as a result of arm swing constriction are limited to walking or running at a constant velocity; however no previous research has been completed involving arm or leg swing constraints during acceleration, but does lead to interesting questions. For example, since large propulsive forces are demanded of the legs during the initiation of acceleration, does arm constriction lead to more trunk rotation? This would be the expected outcome if the arms simply serve as passive dampers. Through recent research, the fascia is now understood to be a powerful force generator (Myers, 2008) and may have an effect on acceleration. Consideration of this tissue and capability of force generation will be pursued by constraining the arms and/or legs. If motion is limited at the distal fascial link, does energy transfer diminish or increase? There is much to be researched involving arm and leg swing constriction during bipedal acceleration. Maximum running speed and acceleration are essential to performing many different individual and team sports. To date, only one study has been conducted that has directly researched arm swing constraints during running, however no previous research has been completed involving arm or leg swing constraints during acceleration. Therefore this study endeavors: (a) to determine the exact role of the lumbopelvic-hip complex during bipedal acceleration; and (b) to determine the overall effects of arm and leg swing constraints on an athlete accelerating by examining kinematics, kinetics, and muscular activity. The outcome of this research will advance the literature on the acceleration phase of sprinting and provide training techniques to athletes and coaches on improving overall performance of sprinting in sporting events as well as expanding the current understanding on the role of the arms and pelvis during locomotion.

CHAPTER III

METHODS

The purpose of this project was to investigate the lumbopelvic-hip complex in bipedal acceleration while examining differences in kinematics, kinetics, and electromyography with different arm and leg constraints. The participants were tested while completing a 10 meter (m) sprint from a standing start position. The participants were instructed to continue running through the 10 m mark while increasing velocity over the entire interval. Participants used the same preferred footwear throughout the experiment.

During the study four randomized conditions were utilized. The first condition consisted of participants sprinting normally with the arms free to move (N). Participants were instructed to complete the initiation of the sprint to achieve maximal velocity. The participants had free range of motion at all joints of the body. The second condition constrained the upper arms of the participant at the humerus (CUA) (Figure 7), allowing for elbow flexion/extension in addition to free range of motion at the pelvis and lower extremities. The humeri of the participants were constrained using an ACE bandage wrap. The third condition constrained arms at the humerus and the forearm (CFA) (Figure 8) while the pelvis and lower extremities had free range of motion. This condition limited motion at the shoulder, elbow and proximal radio-ulnar joints. The shoulders remained adducted by the participants' sides, elbows were extended and the proximal radio-ulnar joints were slightly pronated such that the palms will be facing the body. The fourth condition constrained the participants at the femur (CH) (Figure 8), limiting motion at the hip while the pelvis and upper extremities had full range of motion.

The third chapter presents the methodology used to address the purposes of this study with the following sections: (a) participants and setting, (b) instrumentation, (c) design and procedures, and (d) statistical analysis.

Participants

Fifteen currently competing male rugby athletes were recruited to participate in this research study. In an attempt to maintain propriety, females were excluded and only males were recruited. In order to qualify for participation in this study, individuals must have been between the ages of 19-30 years and not currently have any limitations that would inhibit them from performing repeated, maximal sprint bouts of 10m.

Setting

All testing and data collection occurred in the Sport Biomechanics Laboratory at Auburn University (1127 Beard-Eaves Memorial Coliseum). All equipment and data were held in the Sport Biomechanics Laboratory. The running surface consisted of a flat elevated wooden runway, measuring 15m long by 1.2 m wide, with two embedded AMTI force plates located within the Vicon camera capture volume (Figure 9**)**. In the present study, a distance of 10m was chosen as the outcome of this research is not designed for the 100m race, but for the more generalized use of maximal sprinting for most athletes involved in competitive sports.

Instrumentation

Kinetics:

Ground reaction forces and impulses were examined using 2 AMTI Force OR6-7-1000 Biomechanics Platforms (Advanced Medical Technology Inc., Watertown, Massachusetts) **(**Figure 2**)** and amplifiers (MiniAmp MSA-6, Advanced Medical Technology, Inc. Watertown, Massachusetts) **(**Figure 3**)**. The force plates utilize four force transducers, recorded in Newtons at a sampling rate of 1000 Hz. Two force plates were utilized to ensure complete contact by the foot of the participant while performing the sprint. The force plates were embedded in the runway, approximately 7-8 meters from the start position and recorded ground reaction forces and impulses in Newtons and Newton*seconds, respectively. Force and impulse were collected during the contact phase in the vertical (VGRF and VGRI) and anterior/posterior directions, specifically the braking (BGRF and BGRI) and propulsive phases (PGRF and PGRI).

Figure 2. AMTI OR-6 Force Platform.

Figure 3. AMTI MiniAMP MSA-6.

Three-dimensional Kinematics:

A 10 camera Vicon® MX motion analysis system (Vicon®, Los Angeles, CA, USA) with a sampling frequency of 288 Hz (Devroey et al., 2007)**(**Figure 4) were used to collect three-dimensional kinematic measures of the ankle, knee, hip, pelvic girdle, shoulder, and elbow joints during each testing condition. Thirty- four, 14 mm, spherical retro-reflective markers (MKR-6.4, B & L Engineering, Tustin, California, USA) (Figure 5) were attached to the locations, identified in Figure 8 & Table 2, on each participant's body with double sided tape (CLEAR-1R36, Hair Direct Inc., Bainbridge, Pennsylvania, USA). The standard Plug-In-Gait marker placement protocol was used. Joint position data were calculated by Visual 3D (C-Motion Research Biomechanics, Germantown, Maryland, USA) software using a modified Plug-In-Gait model (Figure 6, Figure 7 and Table 1). Ten Vicon MX cameras were arranged to encapsulate a volume, which was large enough to capture at least two full steps. From this motion capture data joint kinematics in all three planes of motion were determined.

Figure 4. Vicon T-Series Cameras.

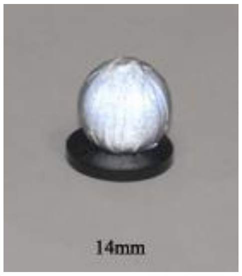

Figure 5. Retro-reflective markers.

Marker Name	Position	Segment
R/LFHD	Right/Left Front of Head	Head
R/LBHD	Right/Left Back of Head	Head
CLAV	Sternoclavicular Joint	Chest
STRN	Sternum	Chest
C7	7th Cervical Vertebrae	Spine
T10	10th Thoracic Vertebrae	Spine
RBAK	Right Midtorso	Asymmetry Identification
R/LSHO	Right/Left Acromioclavicular Joint	Shoulder
R/LELB	Right/Left Lateral Aspect of Humeroulnar Joint	Elbow
R/LWRA	Right/Left Radius	Wrist
R/LWRB	Right/Left Distal End of Ulna	Wrist
R/LFIN	Right/Left 3rd Metacarpophalangeal Joint	Knuckle/Finger
R/LASI	Right/Left Anterior-Superior Iliac Spine	Pelvis
SACR	Sacrum	Pelvis
R/LTHI	Right/Left Lateral Aspect of Femur	Upper Leg
R/LKNE	Right/Left Tibiofemoral Joint	Knee
R/LTIB	Right/Left Lateral Aspect of Fibula	Lower Leg
R/LANK	Right/Left Lateral Malleolus	Ankle
R/LTOE	Right/Left Distal End of 2nd Metatarsal	Toe/Foot
R/LHEE	Right/Left Calcaneus	Heel

Table 1. Vicon Plug-In-Gait marker locations.

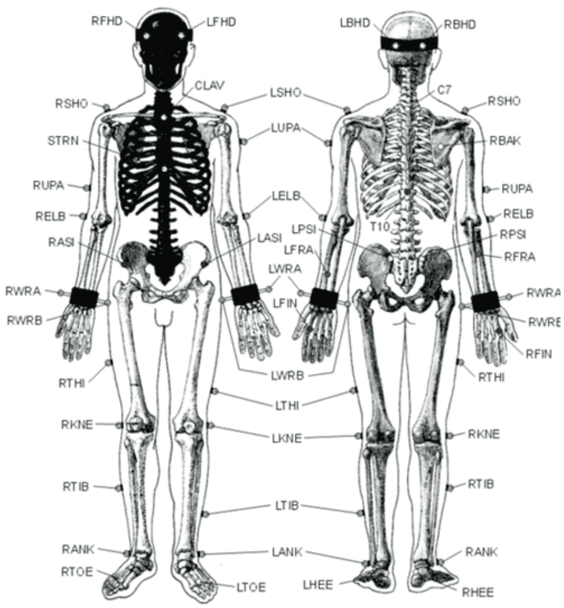

Figure 6. Vicon Plug-In-Gait Model marker locations.

Figure 7. Human Plug-In-Gait Model.

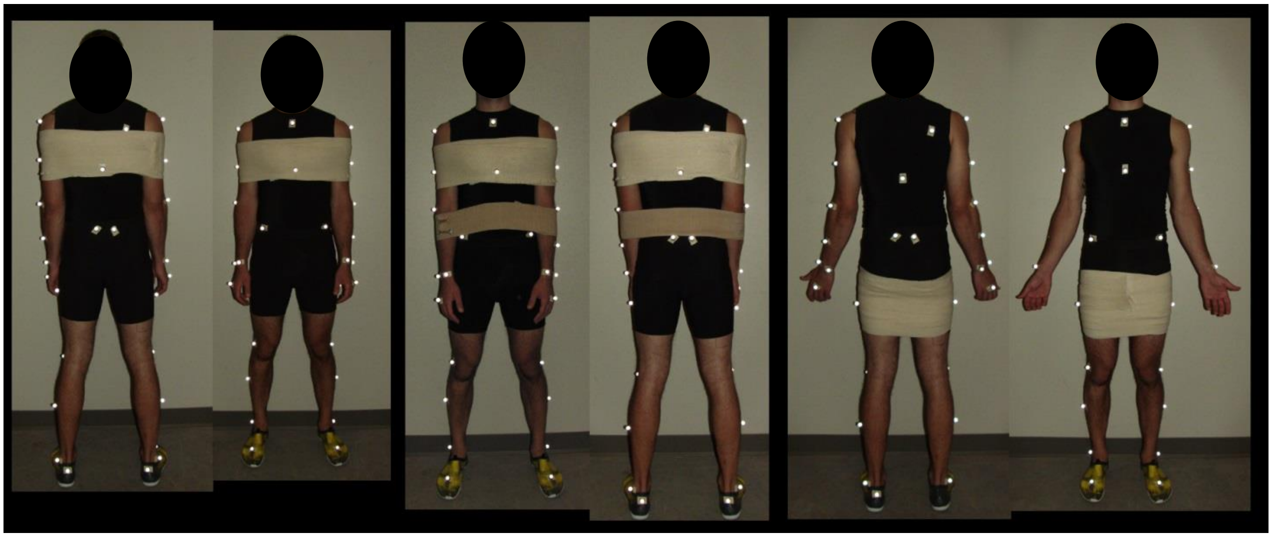

Figure 8. CUA, CFA and CH conditions.

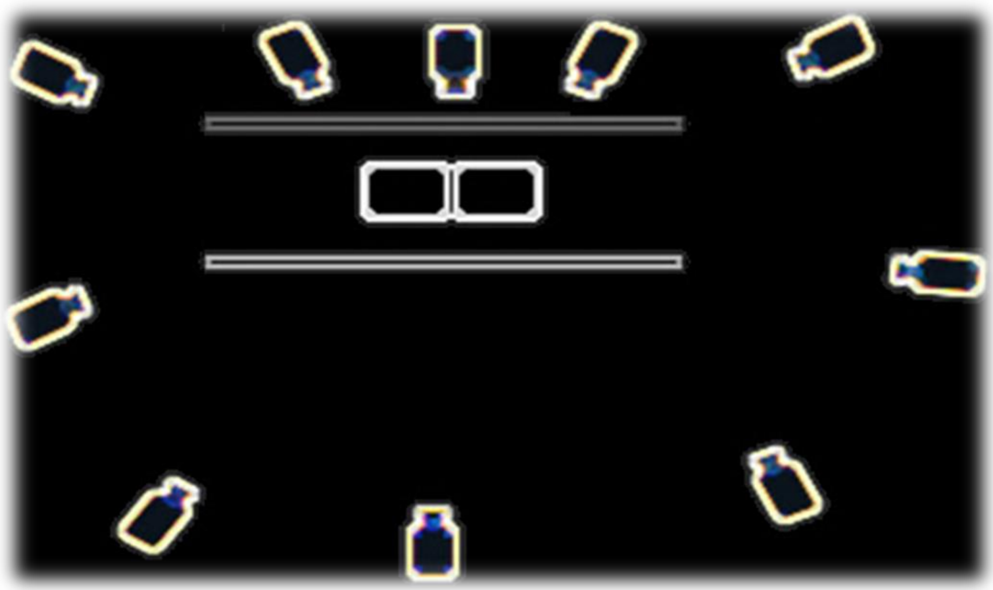

Figure 9. Data collection setup in the Auburn University Sport Biomechanics Laboratory with 2 AMTI force platforms and 10 camera Vicon motion capture setup.

Kinematic measures during initial contact and toe off and peak measured during the flight phase were considered for the following components: Ankle plantar/dorsiflexion (AA), knee flexion/extension (KA), hip flexion/extension (HA), right and left transverse pelvic girdle rotation (RTVPGR, LTVPGR), shoulder angle (SA), and elbow angle (EA). Spatial-temporal parameters including the measures of flight time (FT), contact time (CT), and step length (SL) were analyzed for each participant.

Surface Electromyography (sEMG):

Lower leg muscular activity of the gluteus maximus (GM) and upper extremity muscular activity of the latissimus dorsi (LD) (Figure 12) were collected using Ambu Blue Sensor M-00-S self-adhesive (Ag/AgDl) snap dual electrodes (Ambu, Ballerup, Denmark). The sEMG leads were connected to a Noraxon® Telemyo 2400R-World Wide Telemetry receiver (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) (Figure 10). A Noraxon® Telemyo 2400T V2 wireless transmitter (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA) (Figure 11) was connected to the Telemetry receiver to collect sEMG data. The software utilized for data collection and processing was the MyoResearch XP Master Edition© 1.07.41 (Noraxon® U.S.A. Inc., Scottsdale, AZ, USA). The sEMG was sampled at 1000Hz, and post processing included full wave rectification and a finite impulse response filter (gain set at 1000, bandpass of 10-350Hz, and a notch filer of 59.5-60.5Hz). Peak % MVIC of the muscles involved was analyzed to gain a better understanding of the applications of these muscles during the acceleration phase of a sprint gait cycle. Peak sEMG was analyzed during initial contact (IC) and toe off (TO) of the contact phase. Surface Electromyography, kinematic and kinetic data were synchronized with a Vicon MX Control 64-channel A/D board (Vicon, Los Angeles, CA, USA).

Figure 10. Noraxon Telemyo 2400R – World Wide Telemetry Receiver.

Figure 11. Noraxon Telemyo 2400T V2 wireless transmitter.

Figure 12. Electrode placements of the latissimus dorsi and gluteus maximus muscles.

Design and Procedures

The participants arrived to the Sport Biomechanics Laboratory for two days of testing. Day 1 served as a familiarization day where measurements of the participants were taken and participants completed the sprint bouts with the constraint conditions. The testing session on day 2 lasted approximately 120 minutes from subject preparation to the end of data collection. Each participant was given an Auburn University Institutional Review Board approved informed consent form (Appendix 1A) and health-screening questionnaire (Appendix 1B) to review, complete and sign. The health-screening questionnaire was used to eliminate participants who display any type of lower or upper extremity injuries, which would have prevented them from performing twelve bouts of a

10 meter sprint and/or an allergy to adhesives which was used to collect electromyography data and attach the retro-reflective markers.

Once these procedures were completed and the participant was approved to participate, anthropometric data was taken for each individual. The electrodes were placed parallel to the muscle fibers using techniques described by Basmajian and Deluca (1985). The locations were identified and were shaved, abraded and cleaned using medical alcohol wipes. An inter-electrode distance of 20 mm was selected for surface electrode placement. In addition, an electrode was placed on the C7 vertebral prominens to serve as a ground electrode. Following the application of surface electrodes to the participant, a maximal volitional isometric contraction (MVIC) was recorded using the techniques by Kendal et al. (2005) for manual muscle testing. The MVIC data provided baseline data in which sEMG data were compared and presented as a percentage.

Once the electrodes were placed on the participant, 34 retro-reflective markers were attached to the subject with double sided tape (CLEAR-1R36©, Hair Direct Inc.®, Bainbridge, PA, USA) and based on the modified Vicon Plug-In-Gait model (Figure 6 and Figure 7). The participants wore spandex shirts and shorts to ensure limited marker movement during data collection. A static capture was taken of each participant while standing in the middle of the capture volume on the AMTI force plate. Prior to the actual sprinting the participant warmed-up following a self-selected procedure. Once the participant felt ready to perform the tasks, data collection began.

Procedures

Following warm-up the order of the four conditions (N, CUA, CFA and CH) were randomized. The starting locations were determined during the warm-up to ensure that the right foot contacted the force plate completely. The position was altered within a one-meter range for each participant as needed.

1) Participants were prepared prior to testing based on the condition, which was randomly chosen via random number generator.
2) Participants began from a standing start position then sprinted through the capture volume while kinematic, kinetic, and electromyographic data was collected during the acceleration of each athlete.
3) Participants were allowed a minimum of 2 minutes rest in between each trial to ensure fatigue was not an issue.
4) Participants then completed the previously stated steps for two more successful trials. A successful trial was defined as complete contact with either force plate by the participants' right foot.
5) In between conditions, the participant was prepared for the next condition. Therefore, at this time the participant had his arm/leg constraint modified, either added or removed to follow the purpose of the specified condition.
6) The previous steps were followed until three successful trials had been recorded for each of the four conditions.

There was a period of ten minutes rest in between in each condition where the participant received additional instructions from the researchers in preparation for the next condition. Three successful trials of each condition were collected; the trial with the greatest velocity was utilized for data analysis.

Statistical Data Analysis

All statistical analyses were conducted using SPSS software (version 17.0; SPSS Inc., Chicago, IL, USA) and an alpha level was set a priori at, p≤0.05 ("SPSS for Windows, 2007"). Data was imported to SPSS for analysis once reduction was complete. To investigate the effects of constraint on joint kinematics, gait kinetics, and sEMG across all four conditions, repeated measures ANOVAs were incorporated. The independent variable of constraints had four levels: normal (N), constricted upper arm motion (CUA), constricted full arm motion (CFA), and constricted hip motion (CH). The dependent variables were the following: Peak sEMG of the bilateral LD and GM at initial contact ($ILLD_{IC}$, $CLLD_{IC}$, $ILGM_{IC}$, $CLGM_{IC}$) and toe off ($ILLD_{TO}$, $CLLD_{TO}$, $ILGM_{TO}$, $CLGM_{TO}$); Ground reaction force, time to peak ground reaction force and ground reaction impulse during the contact phase in the vertical ($VGRF_{PEAK}$, VGRI, $VGRF_{TTP}$) and anterior/posterior directions, specifically braking and propulsion ($BGRF_{PEAK}$, $PGRF_{PEAK}$, BGRI, PGRI); step lengths (SL_R, SL_L), contact times (CT_R, CT_L), non-support times (NST_{IC}, NST_{TO}); Ankle angle (AA_{IC}, AA_{TO}), knee angle (KA_{IC}, KA_{TO}), hip angle (HA_{IC}, HA_{TO}), pelvis ($APPGR_{IC}$, $APPGR_{TO}$, $TVPGR_{IC}$, $TVPGR_{TO}$, $LPGR_{IC}$, $LPGR_{TO}$), trunk (TF_{IC}, TF_{TO}, LTF_{IC}, LTF_{TO}, TR_{IC}, TR_{TO}), shoulder angle (SA_{IC}, SA_{TO}), and elbow angle (EA_{IC}, EA_{TO}) at initial contact and toe off.

Post hoc analyses and pairwise comparisons were performed for each dependent variable which exhibited statistical significance. Pairwise comparisons were completed for significant main effects, from the repeated measures ANOVA, between the independent variables.

CHAPTER IV

RESULTS

This study was designed to analyze the kinematics, kinetic and muscle activity during the acceleration phase of a sprint with and without constricted arm and leg motion. Specifically, the purpose of this investigation was addressed in two points: 1) to determine the role of the lumbopelvic-hip complex (LPHC) during the initial acceleration of a sprint and 2) to determine the overall effects of different arm and leg constraints on sprinting mechanics by examining the spatial-temporal and joint position, ground reaction forces and impulses and musculature activity, specifically the latissimus dorsi (LD) and gluteus maximus (GM), at different points during bipedal locomotion acceleration. The following chapter presents the results of the current study and includes the following sections: Section 1: Participant demographics, Section 2: The effect of the normal (N), constrained upper arm (CUA), constricted full arm (CFA) and constricted hip (CH) conditions on spatial-temporal position, Section 3: The effect of the CUA, CFA and CH conditions on joint kinematics, Section 4: The effect of the CUA, CFA and CH conditions on ground reaction forces and impulses, and Section 5: The effect of the CUA, CFA and CH conditions on LD and GM activity.

Section 1: Participant Demographics

Fifteen members of a collegiate level Rugby Club Team volunteered for participation in the current study and met the initial qualifications. Upon answering the medical questionnaire (Appendix A), all volunteers were included in the study. Following completion of the questionnaire, each participant signed an Informed Consent form which was approved by the Auburn University Institutional Review Board for Human Subjects (Appendix B). Table 2 summarizes the participant demographics averages.

Table 2.
Participant Demographics

	Mean	Standard Deviation
Age (years)	22	± 2.8
Height (m)	1.82	± .06
Weight (kg)	77.95	± 10

Section 2: Spatial-Temporal Kinematics

In an attempt to understand the effects of arm and leg swing constraints (CUA, CFA and CH) during bipedal acceleration, the spatial-temporal kinematics, specifically, step lengths (SL_R, SL_L), non-support times (NST_R, NST_L) and contact times ($CT_{R,}$ CT_L) were investigated. These data were compared to a normal condition (N).

In order to determine the influence of constraint on right step length, a 1 (right step length, SL_R) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the spatial-temporal kinematics of right step length (SL_R) ($F(1,14) = .099$, $p = 0.758$, $\eta^2 = 0.007$, *Power* = 0.060). Although there was no significant difference between the four constraint conditions, the N condition yielded a larger SL_R indicating a longer during the non-constrained condition compared to the constrained conditions (CUA, CFA and CH) (Figure 13).

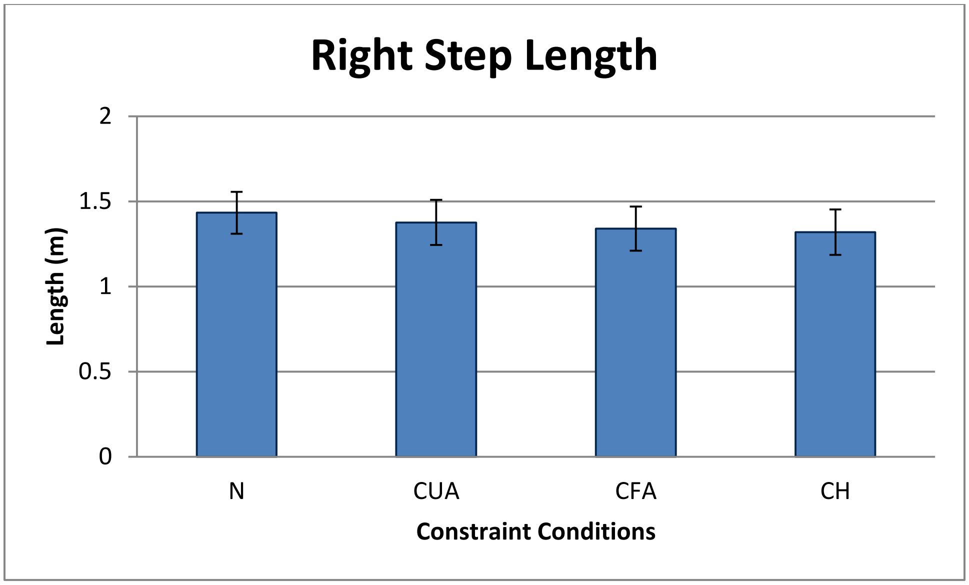

Figure 13. Effects of constraint on right step length.

A separate analysis was conducted to determine the influence of arm and leg constraint on left step length. A 1 (left step length, SL_L) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the spatial-temporal kinematics of left step length (SL_L) ($F(1,14) = 9.553$, $p = 0.000$, $\eta^2 = 0.406$, *Power* = 0.995). This indicates that different constraint conditions altered left step length. Follow-up pairwise comparisons for SL_L were conducted and significant differences were noted between the following conditions: N and CUA ($p = .038$), N and CFA ($p = .001$), N and CH ($p = .002$), CUA and CFA ($p = .002$), and CUA and CH ($p = .040$) (Figure 14). The results indicate that the full constrained conditions of arms and hips, CFA and CH yielded significantly smaller strides than the CUA and N conditions.

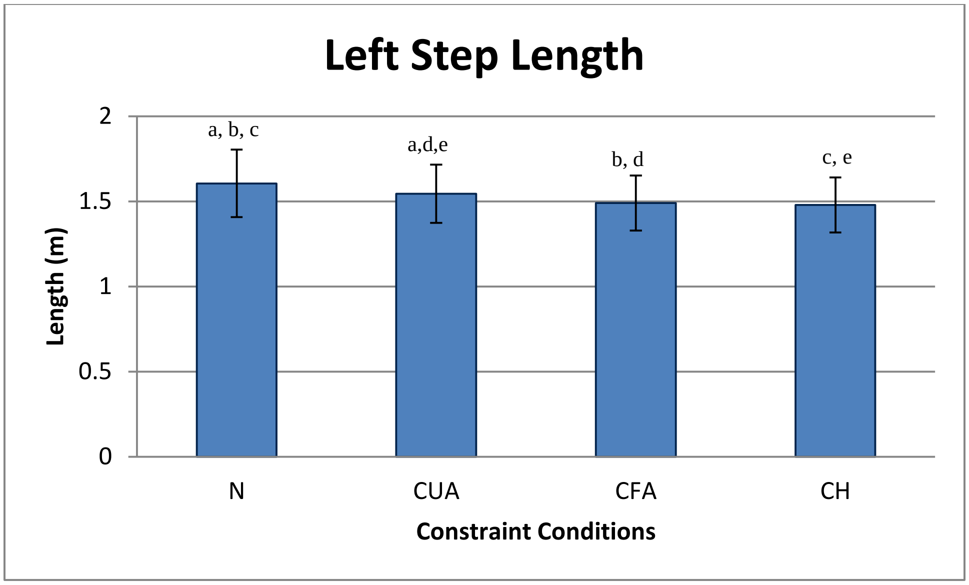

Figure 14. Effects of constraint on left step length.
N-CUA, N-CFA, N-CH, CUA-CFA, CUA-CH: [a,b,c,d,e] p <.05.

While investigating the influence of constraints on non-support time prior to right foot contact, a 1 (non-support time, NST_R) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the spatial-temporal kinematics of non-support time prior to right foot contact (NST_R) ($F(1,14) = 3.584$, $p = 0.079$, $\eta^2 = 0.204$, *Power* = 0.422). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger NST_R indicating a longer step during the CFA condition when compared to N, CUA and CH (Figure 15).

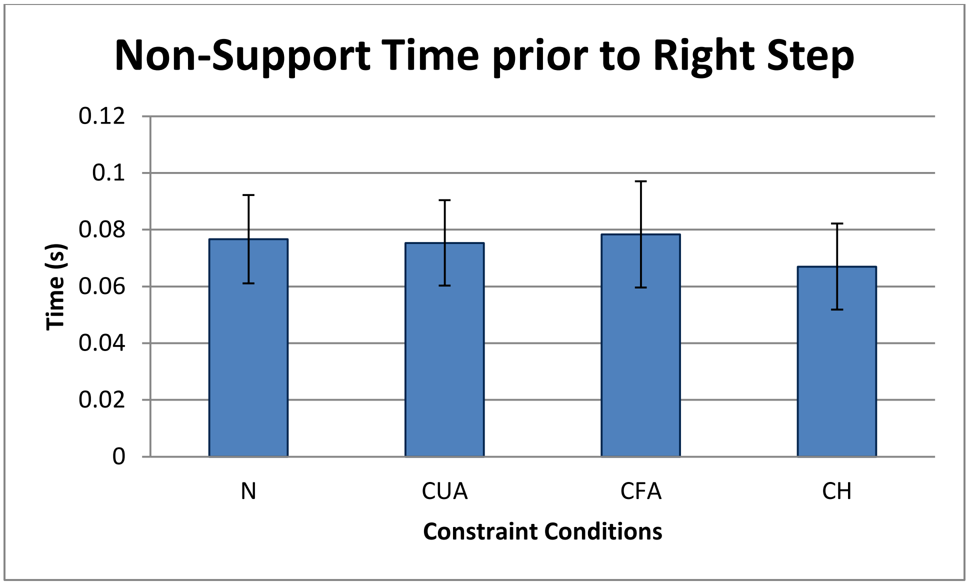

Figure 15. Effects of constraint on non-support time prior to right foot contact.

In addition, to investigate the influence of constraint on non-support time prior to left foot contact, a 1 (non-support time, NST_L) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the spatial-temporal kinematics of non-support time prior to left foot contact (NST_L) ($F(1,14) = 1.585$, $p = 0.229$, $\eta^2 = 0.102$, *Power* = 0.217). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger NST_L indicating a longer step during CFA than N, CUA and CH (Figure 16).

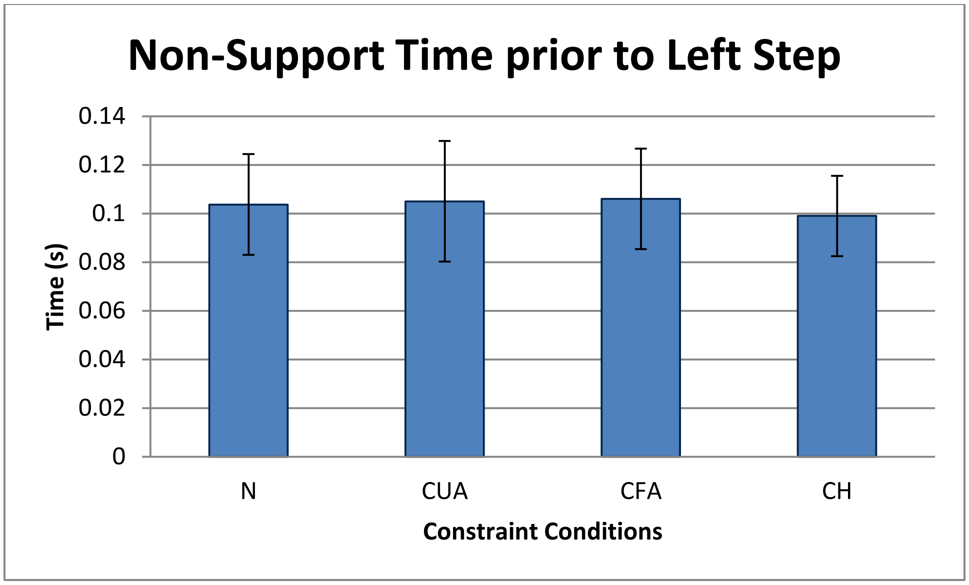

Figure 16. Effects of constraint on non-support time prior to left foot contact.

To investigate the influence of constraint on the contact time during right foot contact, a 1 (contact time, CT_R) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the spatial-temporal kinematics of contact time during right foot contact (CT_R) ($F(1,14) = 3.3586$, $p = 0.079$, $\eta^2 = 0.204$, *Power* = 0.422). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger CT_R indicating longer time in contact with the ground during the constrained condition. In addition, contact time for condition N was the lowest across all conditions, CUA, CFA and CH (Figure 17).

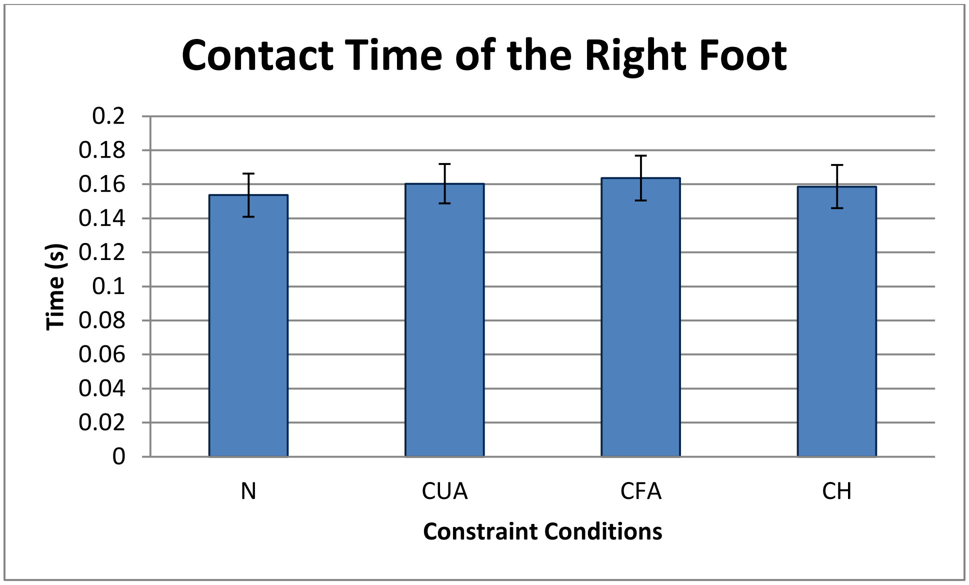

Figure 17. Effects of constraint on contact time during right foot contact.

In addition, to determine the influence of constraint on the contact time during left foot contact, a 1 (contact time, CT_L) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the spatial-temporal kinematics of contact time during left foot contact (CT_L) ($F(1,14) = 57.11$, $p = 0.000$, $\eta^2 = 0.803$, *Power* = 1.000). Follow-up pairwise comparisons for CT_L were conducted and significant differences were noted between the following conditions: N and CUA (p =.002), N and CFA (p = .000), N and CH (p = .000), and CUA and CFA (p = .033) (Figure 18). Results from the statistical analysis demonstrate the contact time of the left foot significantly increased with greater constraints on the athlete. The contact time of the left side was significantly smaller in N condition than the other three conditions, CUA, CFA and CH.

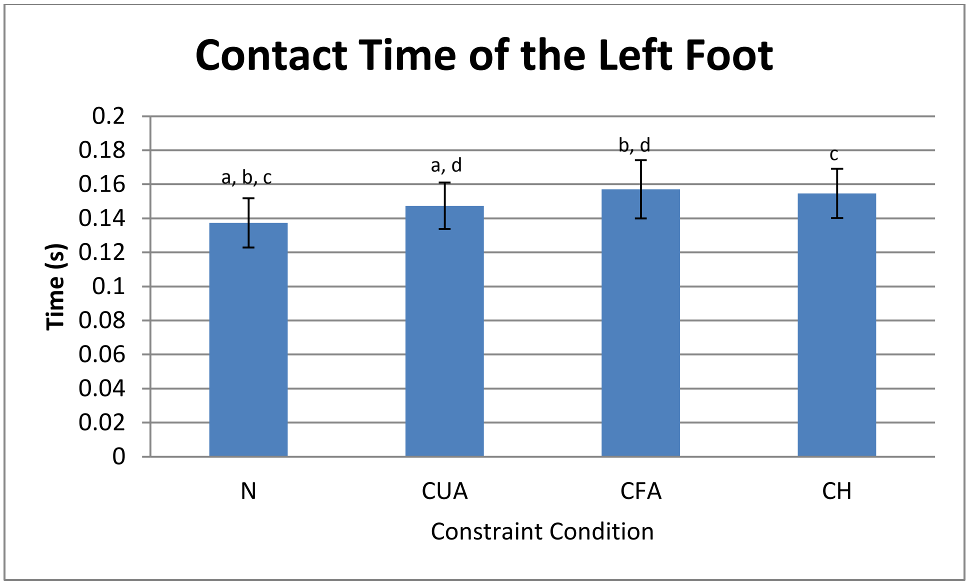

Figure 18. Effects of constraint on contact time during left foot contact.
N-CUA, N-CFA, N-CH, CUA-CFA: [a,b,c,d] p<.05.

In order to determine the effect of constraint on velocity during steps 4 and 5, a 1 (velocity, VEL) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on velocity (VEL) ($F(1,14) = 21.61$, $p = 0.000$, $\eta^2 = 0.607$, *Power* = 0.991). This indicates that different constraint conditions altered velocity during steps 4 and 5. Follow-up pairwise comparisons for VEL were conducted and significant differences were noted between the following conditions: N and CUA (p =.005), N and CFA (p = .000), N and CH (p = .002), and CUA and CFA (p = .002) (Figure 19). Results from the statistical analysis demonstrate the velocity significantly decreased with greater constraints on the athlete. The velocity was significantly greater in the N condition versus the three other conditions, CUA, CFA and CH.

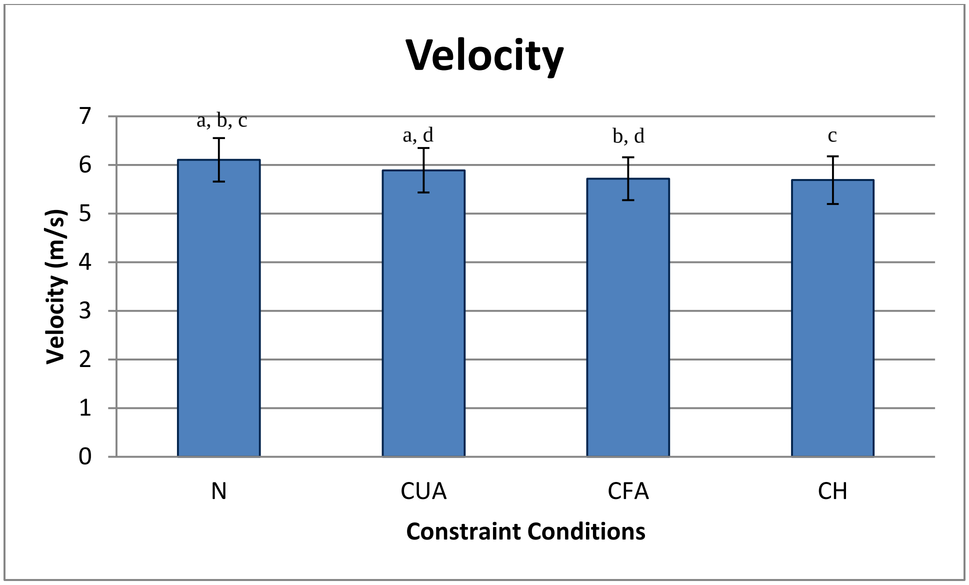

Figure 19. Effects of constraint on velocity during steps 4 and 5.
N-CUA, N-CFA, N-CH, CUA-CFA: [a,b,c,d] p<.05.

Section 3: Joint Kinematics

In addition to investigating the effects of constraint on spatial-temporal kinematics, joint kinematics at the ipsilateral (IL) ankle (AA_{IC}, AA_{TO}), IL knee (KA_{IC}, KA_{TO}), IL hip (HA_{IC}, HA_{TO}), pelvis ($APPGR_{IC}$, $APPGR_{RO}$, $TVPGR_{IC}$, $TVPGR_{TO}$, $LPGR_{IC}$, $LPGR_{TO}$), trunk (TF_{IC}, TF_{TO}, LTF_{IC}, LTF_{TO}, TR_{IC}, TR_{TO}), IL shoulder (SA_{IC}, SA_{TO}) and IL elbow (EA_{IC}, EA_{TO}) were also investigated during initial contact and toe off of the fourth step of the right foot. All joint kinematics data were normalized to the anatomical neutral position of the participants.

In order to determine the effect of constraint on ankle angle at initial contact of the right foot, a 1 (ankle angle, AA_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of ankle angle at initial contact of the right foot (AA_{IC}) (F(1,14)

= 4.4176, p = 0.060, η^2 = 0.230, *Power* = 0.477). Data are represented in degrees and represents the angle between the foot and the shank of the participants. Measures smaller than 90 degrees implies dorsiflexion and those larger than 90 degrees represent plantarflexion. Although there was no significant difference between the four constraint conditions, the CUA and N conditions yielded smaller AA_{IC} indicating greater dorsiflexion than the CFA and CH conditions (Figure 20).

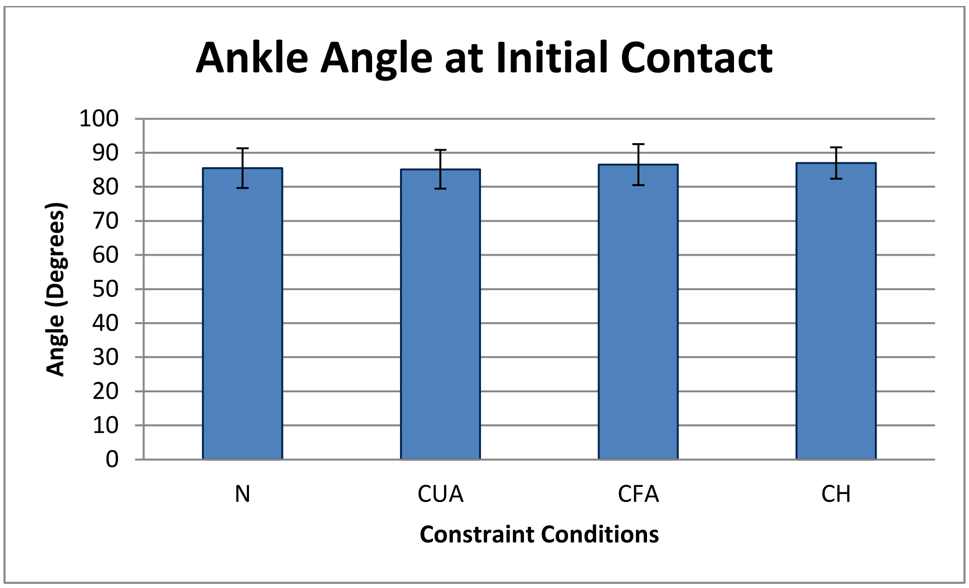

Figure 20. Effects of constraint on ankle angle at initial contact.

Additionally, to determine the influence of constraint on ankle angle at toe off, a 1 (ankle angle, AA_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a p value set apriori at p <0.05. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of ankle angle (AA_{TO}) (F(1,14) = .001, p = 0.979, η^2 = 0.000, *Power* = 0.05). Data are represented in degrees and represents the angle between the foot and the shank of the participants. Measures smaller than 90 degrees implies dorsiflexion and those larger than 90 degrees represent plantarflexion. Although there was no significant difference between the four constraint conditions, the CUA condition yielded a larger AA_{TO} indicating greater plantar flexion during the upper arm constrained condition (Figure 21).

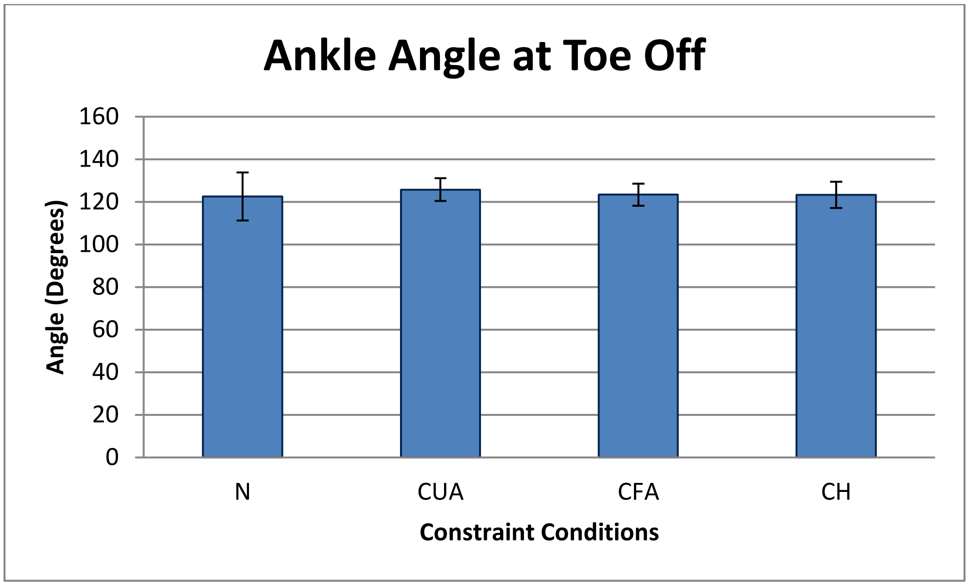

Figure 21. Effects of constraint on ankle angle at toe off of the right foot.

To determine effect of constraint on knee angle at initial contact, a 1 (knee angle, KA_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of knee angle at initial contact (KA_{IC}) ($F(1,14) = .000$, $p = 0.993$, $\eta^2 = 0.000$, *Power* = 0.050). The knee angle data represents the angle between the thigh and shank segments. Although there was no significant difference between the four constraint conditions, the CUA condition yielded a smaller knee angle at initial contact indicating greater knee flexion at initial contact during the constrained condition than in the N, CFA and CH conditions (Figure 22).

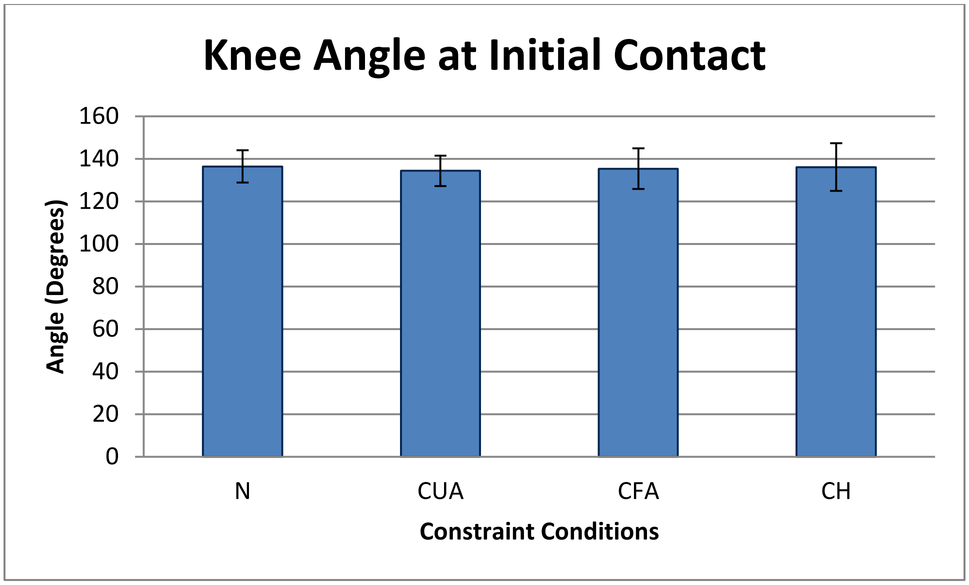

Figure 22. Effects of constraint on knee angle at initial contact of the right foot.

In addition, to determine the influence of constraint on knee angle at toe off, a 1 (knee angle, KA_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of knee angle (KA_{TO}) ($F(1,14) = 9.306$, $p = 0.009$, $\eta^2 = 0.399$, *Power* = 0.810). This indicates that different constraint conditions altered knee angle at toe off. Follow-up pairwise comparisons for KA_{TO} were conducted and significant differences were noted between the following conditions: N and CH (p = .015), CUA and CH (p = .008), and CFA and CH (p = .028) (Figure 23). The knee angle data represents the angle between the thigh and shank segments. The results from this statistical analyses indicate that knee angle at toe off during the CH condition is significantly smaller than in the N, CUA and CFA conditions.

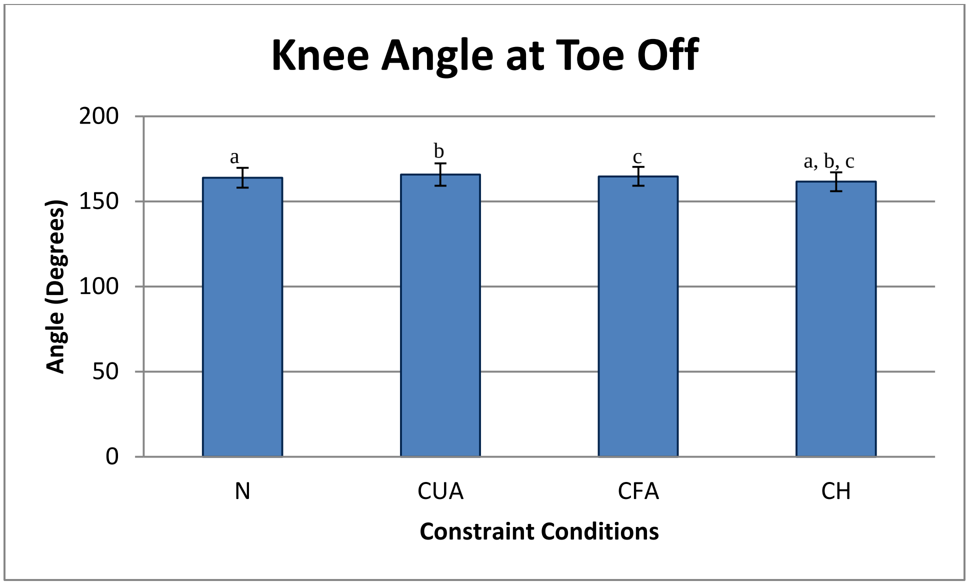

Figure 23. Effects of constraint on knee angle at toe off.
N-CH, CUA-CH, CFA-CH: ^{a,b,c} $p<.05$.

Joint kinematic data at the hip joint were taken with reference to the pelvis measured in degrees. The hip angle was defined as the angle between the femur and the pelvis. To analyze the effects of constraint on hip angle at initial contact, a 1 (hip angle, HA_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of hip angle at initial contact (HA_{IC}) ($F(1,14) = .159$, $p = 0.696$, $\eta^2 = 0.011$, *Power* = 0.066). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger HA_{IC} indicating greater hip flexion at initial contact during the constrained condition than in N, CUA and CH (Figure 24).

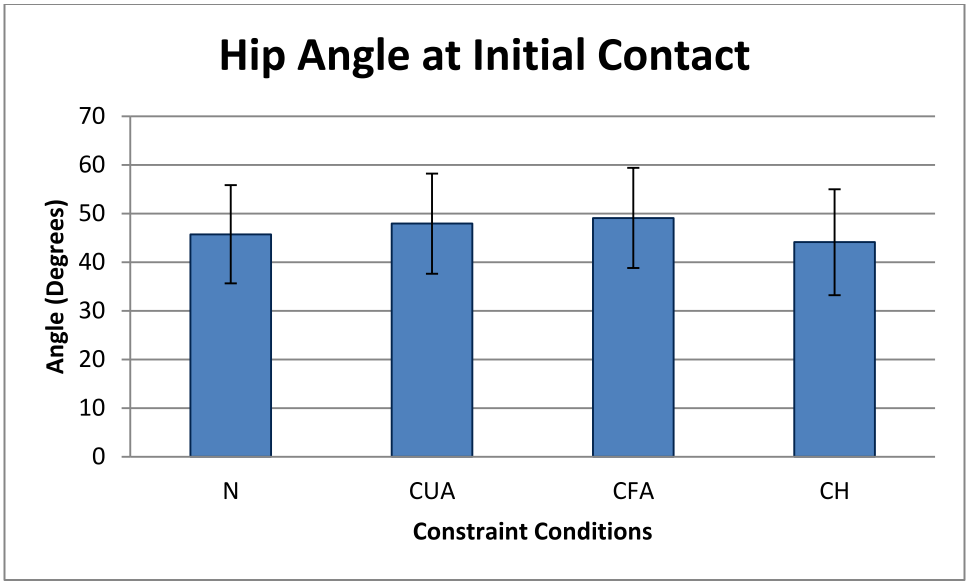

Figure 24. Effects of constraint on hip angle at initial contact.

To determine the influence of constraint on hip angle at toe off a 1 (hip angle, HA_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the hip angle (HA_{TO}) ($F(1,14) = 8.202$, $p = 0.013$, $\eta^2 = 0.369$, *Power* = 0.759). This indicates that different constraint conditions altered hip angle at toe off. Follow-up pairwise comparisons for HA_{TO} were conducted and significant differences were noted between the following conditions: N and CH (p = .004), CUA and CH (p = .031), and CFA and CH (p = .034) (Figure 25). These results indicate that hip angle at toe off is significantly lower in the CH condition when compared to the N, CUA and CFA conditions. Significantly smaller hip angle during the CH condition indicates that the femur was not as extended as when compared to the greater hip angles of the N, CUA, and CFA conditions. Interestingly, CFA indicated the greatest hip extension to occur at toe off between all conditions.

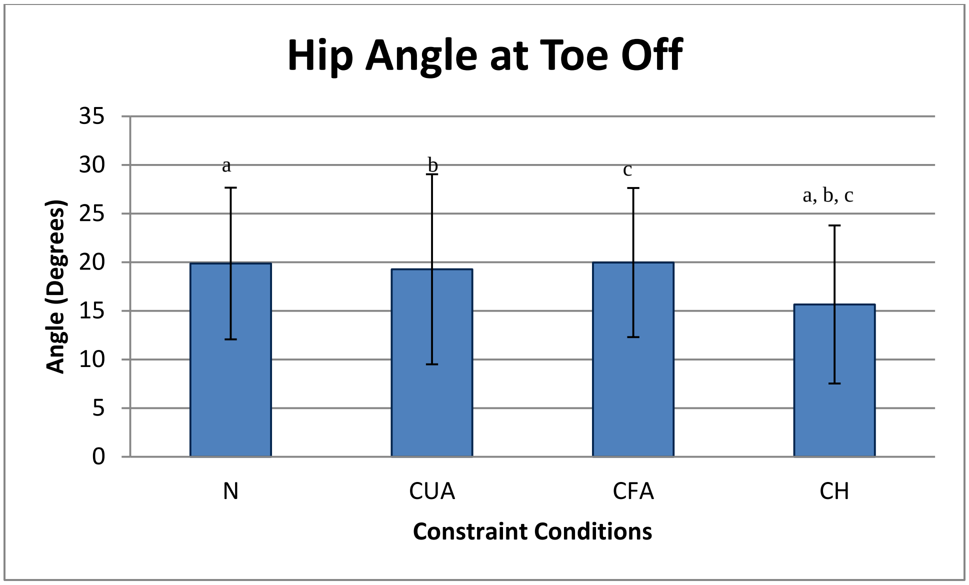

Figure 25. Effects of constraint on hip angle at toe off.
N-CH, CUA-CH, CFA-CH: [a,b,c] p<.05.

In consideration of the pelvis, six motions were measured and statistically analyzed. Pelvic kinematic data were taken in reference to the trunk of the participant. To determine the influence of constraint on pelvic girdle angle in the sagittal plane at initial contact, a 1 (pelvic girdle angle, $APPGR_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of pelvic girdle angle in the sagittal plane at initial contact ($APPGR_{IC}$) ($F(1,14) = 4.646$, $p = 0.049$, $\eta^2 = 0.249$, *Power* = 0.519). This indicates that different constraint conditions altered pelvic girdle angle in the sagittal plane at initial contact. Follow-up pairwise comparisons for $APPGR_{IC}$ were conducted and significant differences were noted between the following conditions: N and CUA (p = .030) and N and CH (p = .006) (Figure 26). The results from this test indicate that the N condition was accompanied with the greatest anterior pelvic girdle rotation when statistically compared to the CUA and CH conditions.

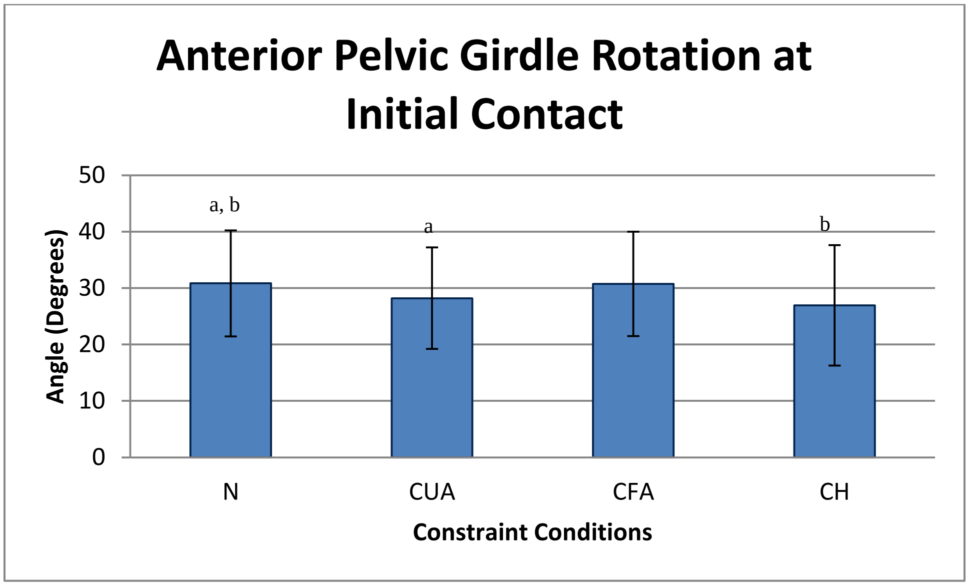

Figure 26. Effects of constraint on pelvic girdle angle in the sagittal plane at initial contact.
N-CUA, N-CH: [a,b] p<.05.

In order to determine the influence of constraint on pelvic girdle angle in the sagittal plane at toe off, a 1 (pelvic girdle angle, $APPGR_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of pelvic girdle angle in the sagittal plane at toe off ($APPGR_{TO}$) ($F(1,14) = 1.070$ $p = 0.319$, $\eta^2 = 0.071$, *Power* = 0.161). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger $APPGR_{TO}$ indicating the greatest anterior pelvic girdle rotation angle during the constrained condition when compared to N, CUA, and CH (Figure 27).

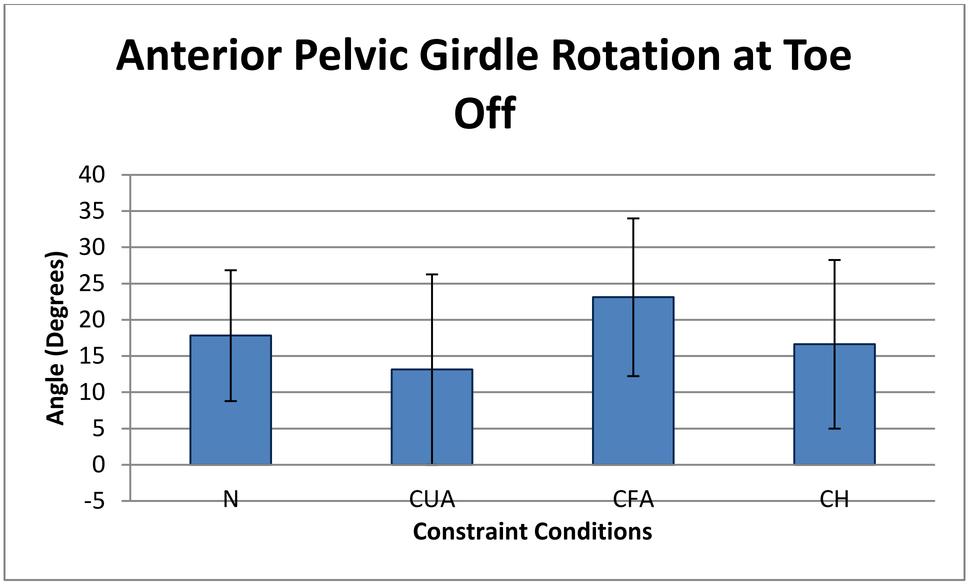

Figure 27. Effects of constraint on pelvic girdle angle in the sagittal plane at toe off.

In order to determine the influence of constraint on pelvic girdle angle in the transverse plane at initial contact, a 1 (pelvic girdle angle, $TVPGR_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the pelvic girdle angle in the transverse plane at initial contact ($TVPGR_{IC}$) ($F(1,14) = .846$, $p = 0.373$, $\eta^2 = 0.057$, *Power* = 0.138). Although there was no significant difference between the four constraint conditions, the N condition yielded a larger $TVPGR_{IC}$ indicating the greatest contralateral (CL) transverse pelvic girdle rotation at initial contact during the non-constrained condition in comparison to CUA, CFA, and CH (Figure 28).

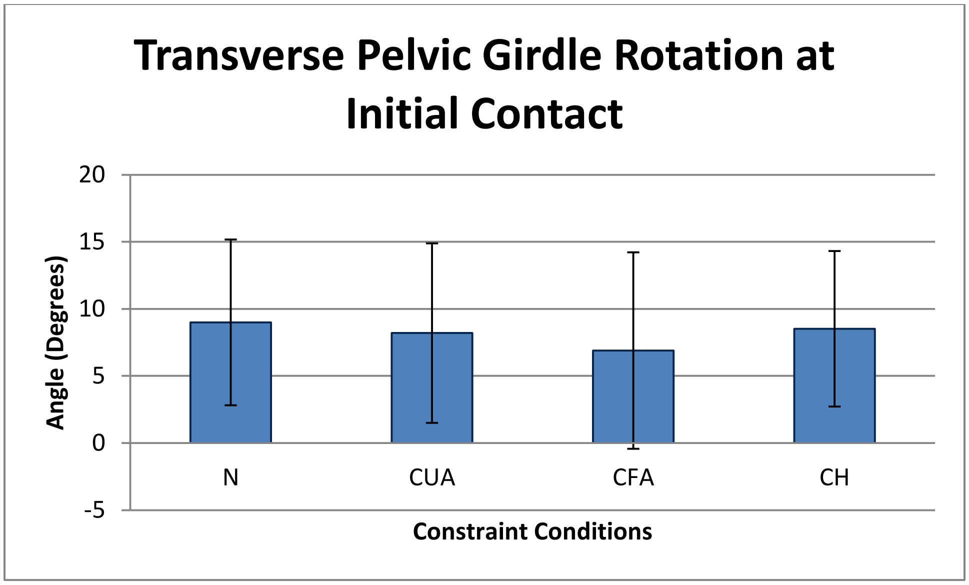

Figure 28. Effects of constraint on pelvic girdle angle in the transverse plane at initial contact.

In order to determine the influence of constraint on pelvic girdle angle in the transverse plane at toe off, a 1 (pelvic girdle angle, $TVPGR_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of pelvic girdle angle in the transverse plane at toe off ($TVPGR_{TO}$) ($F(1,14) = 2.729$, $p = 0.121$, $\eta^2 = 0.163$, *Power* = 0.337). Although there was no significant difference between the four constraint conditions, the CUA condition yielded a larger $TVPGR_{TO}$ indicating the greatest IL transverse pelvic girdle rotation at toe off during the constrained condition when compared to N, CUA, and CH (Figure 29).

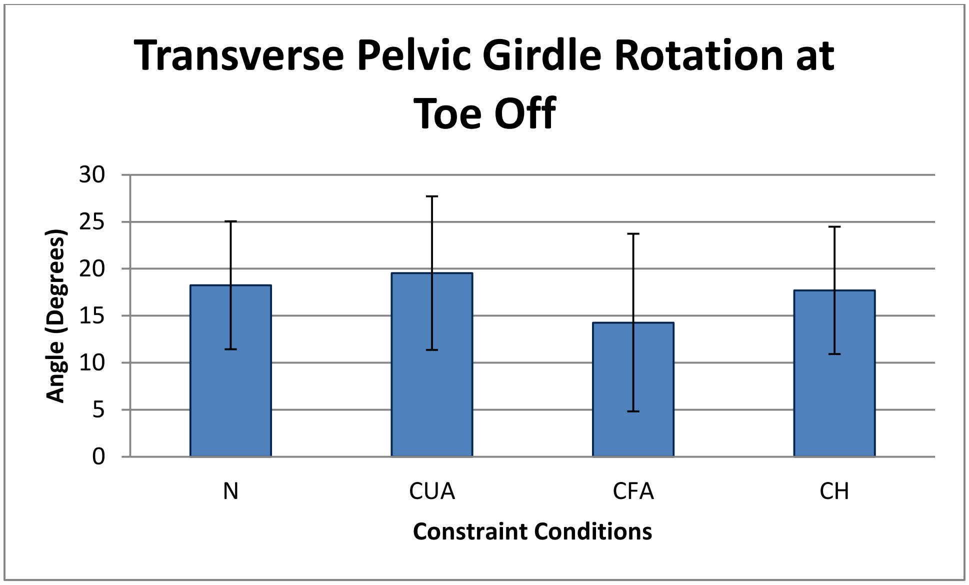

Figure 29. Effects of constraint on pelvic girdle angle in the transverse plane at toe off.

In order to investigate the effects of constraint on pelvic girdle angle in the frontal plane at initial contact, a 1 (pelvic girdle angle, $LPGR_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of the pelvic girdle in the frontal plane at initial contact ($LPGR_{IC}$) ($F(1,14) = 4.658$, $p = 0.049$, $\eta^2 = 0.250$, *Power* = 0.520). This indicates that different constraint conditions altered pelvic girdle angle in the frontal plane at initial contact. Follow-up pairwise comparisons for $LPGR_{IC}$ were conducted and significant differences were noted between the following conditions: N and CFA (p = .005) and CFA and CH (p = .019) (Figure 30). Results from the statistical analysis indicate that the CFA condition had significantly lower IL lateral pelvic girdle rotation when compared to the N and CH conditions.

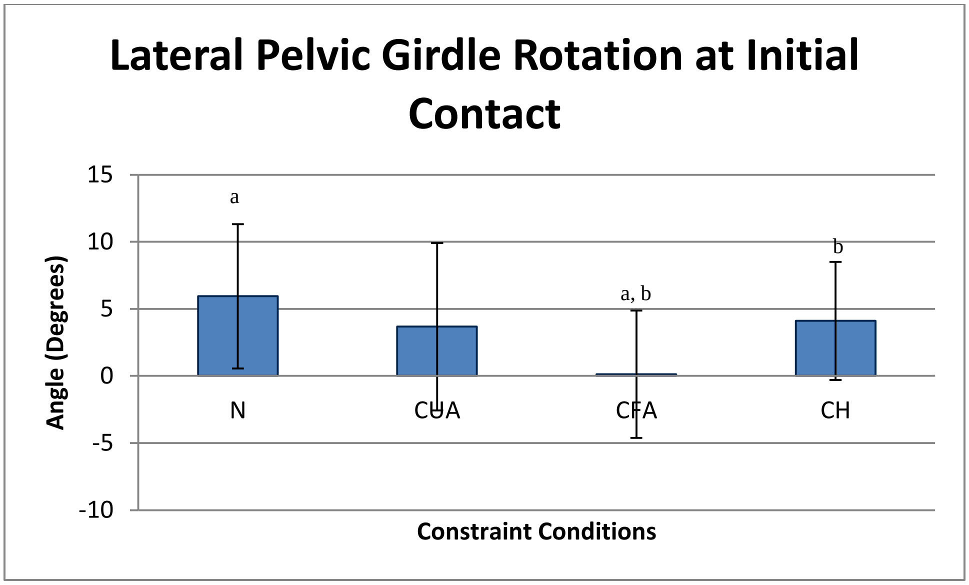

Figure 30. Effects of constraint on pelvic girdle angle in the frontal plane at initial contact.
N-CFA, CFA-CH: [a,b] p<.05.

In order to determine the influence of constraint on pelvic girdle angle in the frontal plane at toe off, a 1 (pelvic girdle angle, $LPGR_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of pelvic girdle angle in the frontal plane at toe off ($LPGR_{TO}$) ($F(1,14) = 3.007$, $p = 0.105$, $\eta^2 = 0.177$, *Power* = 0.365). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger $LPGRF_{TO}$ indicating greater CL lateral pelvic girdle rotation during the constrained condition when compared to N, CUA, and CH (Figure 31).

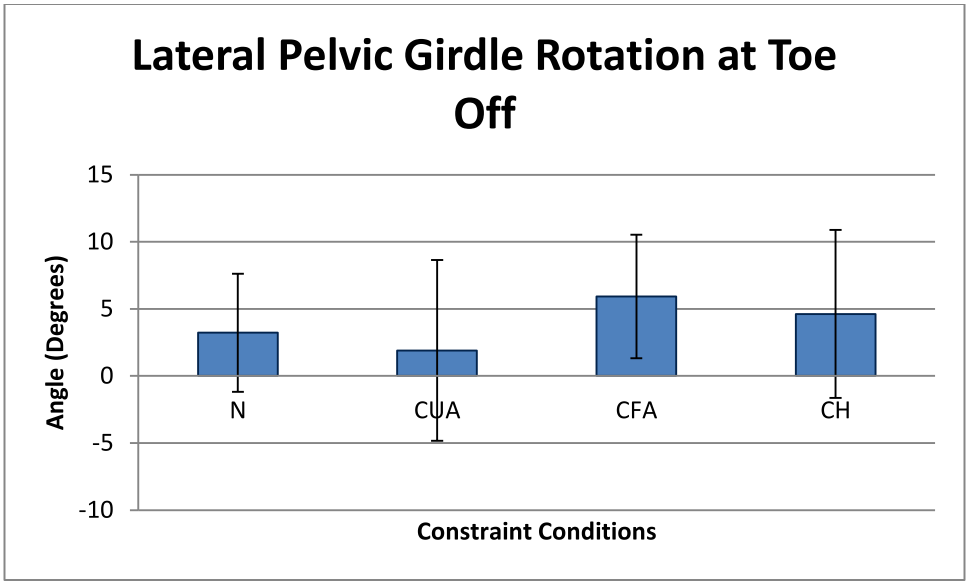

Figure 31. Effects of constraint on pelvic girdle angle in the frontal plane at toe off.

The trunk was the next segment that was analyzed to determine effects between conditions. All trunk data were referenced to the position of the pelvis within each subject. To determine the influence of constraint on trunk angle in the sagittal plane at initial contact, a 1 (trunk angle, TF_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a p value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of trunk angle in the sagittal plane at initial contact (TF_{IC}) ($F(1,14) = 2.1666$, $p = 0.163$, $\eta^2 = 0.134$, *Power* = 0.279). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger TF_{IC} indicating the greatest trunk flexion at initial contact during the constrained condition in comparison to N, CUA and CH (Figure 32).

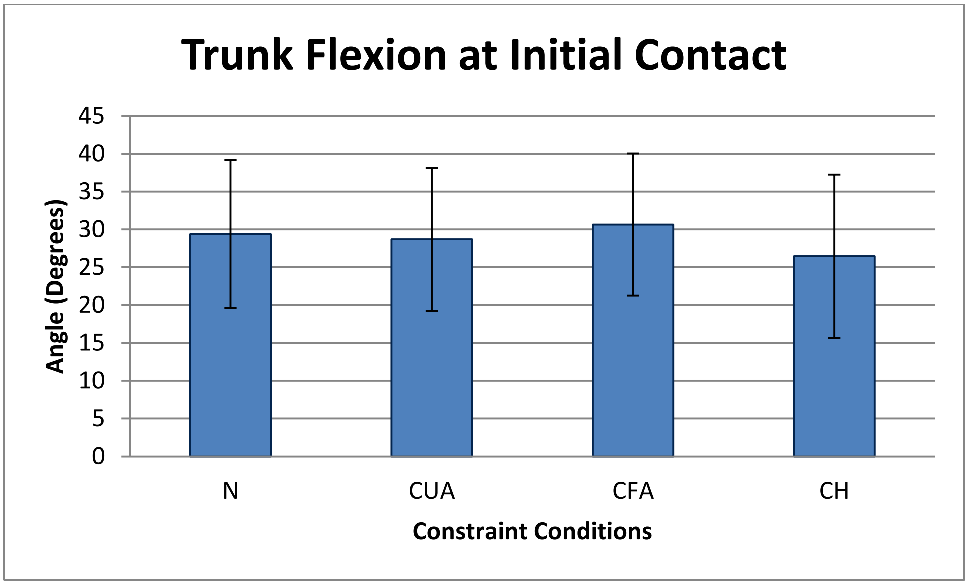

Figure 32. Effects of constraint on trunk angle in the sagittal plane at initial contact.

In order to determine the influence of constraint on trunk angle in the sagittal plane at toe off, a 1 (trunk angle, TF_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of trunk angle in the sagittal plane at toe off (TF_{TO}) ($F(1,14) = .006$, $p = 0.938$, $\eta^2 = 0.000$, *Power* = 0.051). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger TF_{TO} indicating greater trunk flexion at toe off during the constrained condition followed by N, CUA and CH (Figure 33).

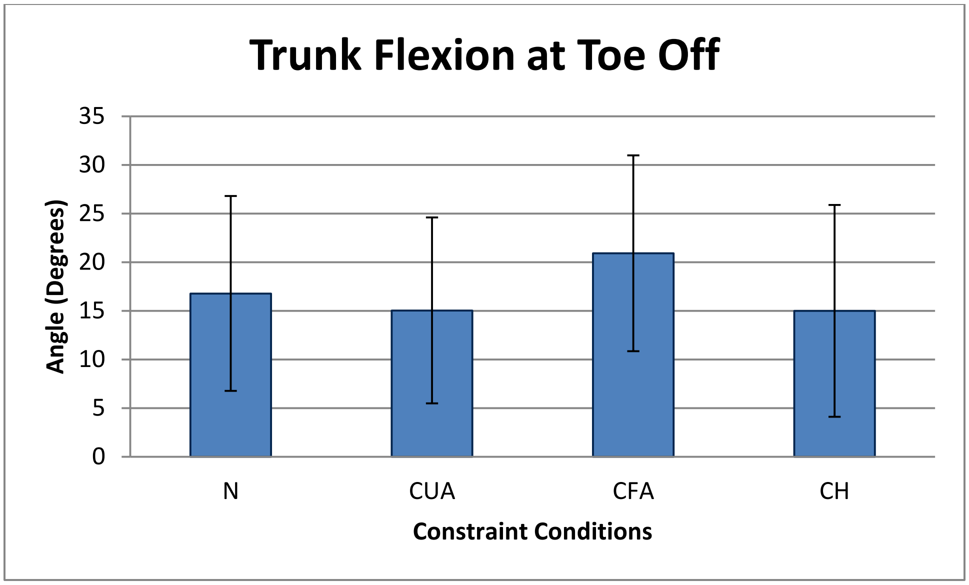

Figure 33. Effects of constraint on trunk angle in the sagittal plane at toe off.

In addition, a 1 (trunk angle, LTF_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$ to determine the effects of constraint on trunk angle in the frontal plane at initial contact. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of trunk angle in the frontal plane at initial contact (LTF_{IC}) ($F(1,14) = 1.720$, $p = 0.211$, $\eta^2 = 0.109$, $Power = 0.231$). Although there was no significant difference between the four constraint conditions, the N condition yielded a larger LTF_{IC} indicating greater IL lateral trunk flexion at initial contact during the non-constrained condition followed by CUA, CFA and CH (Figure 34).

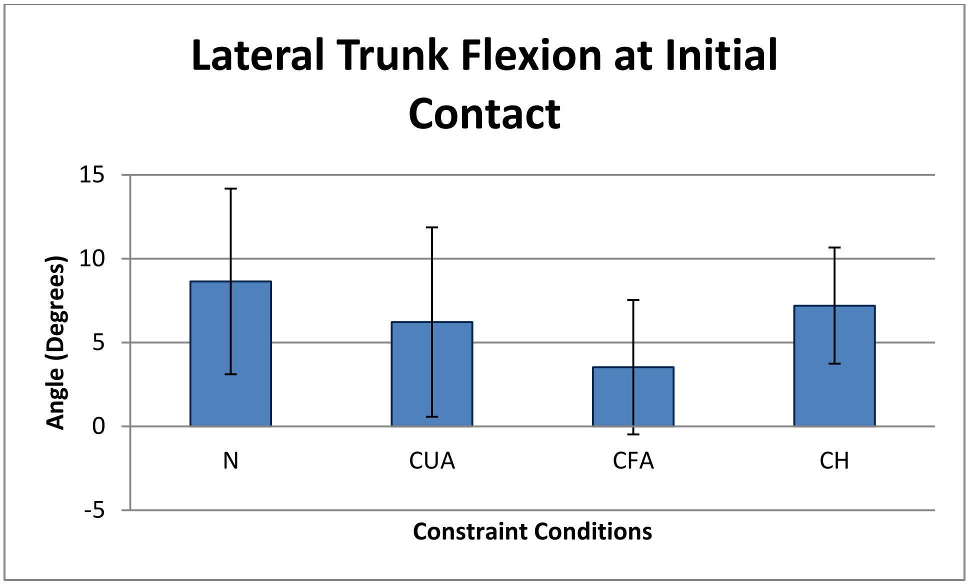

Figure 34. Effects of constraint on trunk angle in the frontal plane at initial contact.

Moreover, to determine the influence of constraint on trunk angle in the frontal plane at toe off, a 1 (trunk angle, LTF_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of trunk angle in the frontal plane at toe off (LTF_{TO}) ($F(1,14) = 2.966$, $p = 0.107$, $\eta^2 = 0.175$, *Power* = 0.361). Although there was no significant difference between the four constraint conditions, the CFA condition yielded a larger LTF_{TO} indicating a greater CL lateral trunk flexion at toe off during the constrained condition followed by CH, N, and CUA (Figure 35).

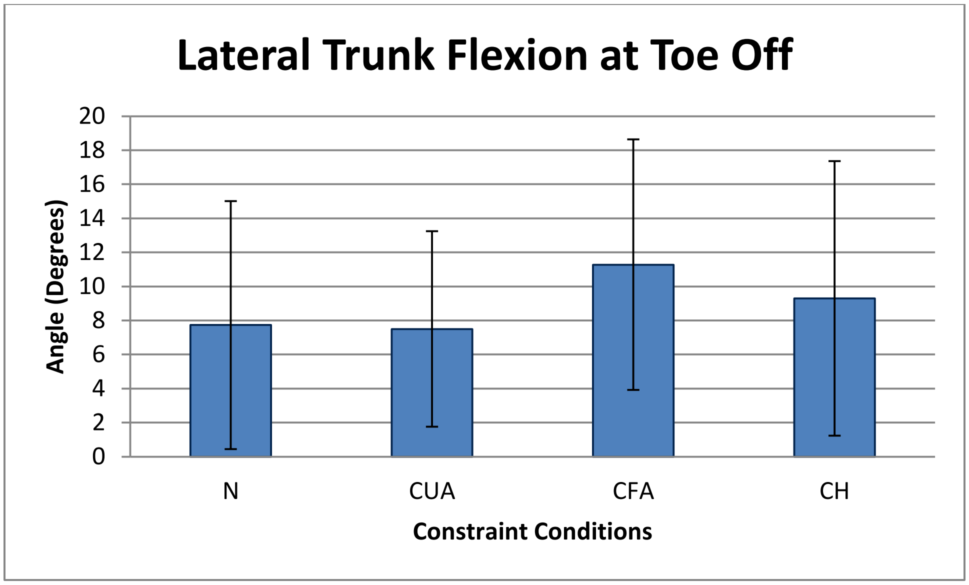

Figure 35. Effects of constraint on trunk angle in the frontal plane at toe off.

Furthermore, a 1 (trunk angle, TR_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$ to determine the influence of constraint on trunk angle in the transverse plane at initial contact. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of trunk angle in the transverse plane at initial contact (TR_{IC}) ($F(1,14) = 3.159$, $p = 0.097$, $\eta^2 = 0.184$, *Power* = 0.381). Although there was no significant difference between the four constraint conditions, the CUA condition yielded a larger TR_{IC} indicating greater IL trunk rotation at initial contact during the constrained condition when compared to N, CUA and CH (Figure 36).

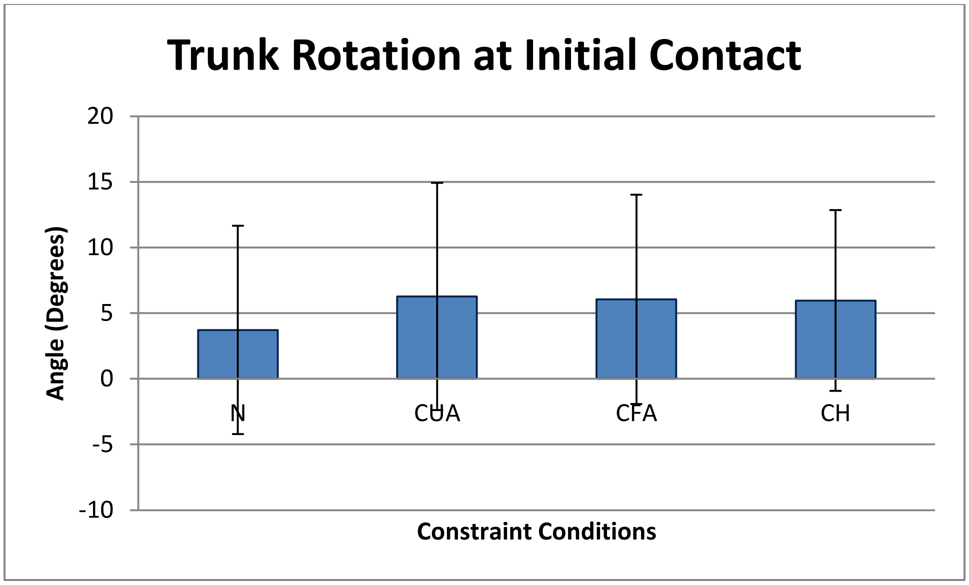

Figure 36. Effects of constraint on trunk angle in the transverse plane at initial contact.

In addition, to determine the influence of constraint on trunk angle in the transverse plane at toe off, a 1 (trunk angle, TR_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of trunk angle in the transverse plane at toe off (TR_{TO}) ($F(1,14) = 1.514$, $p = 0.239$, $\eta^2 = 0.098$, *Power* = 0.209). Although there was no significant difference between the constraint conditions, the CUA condition yielded a larger TR_{TO} indicating higher CL trunk rotation at toe off when compared to N, CUA and CH (Figure 37).

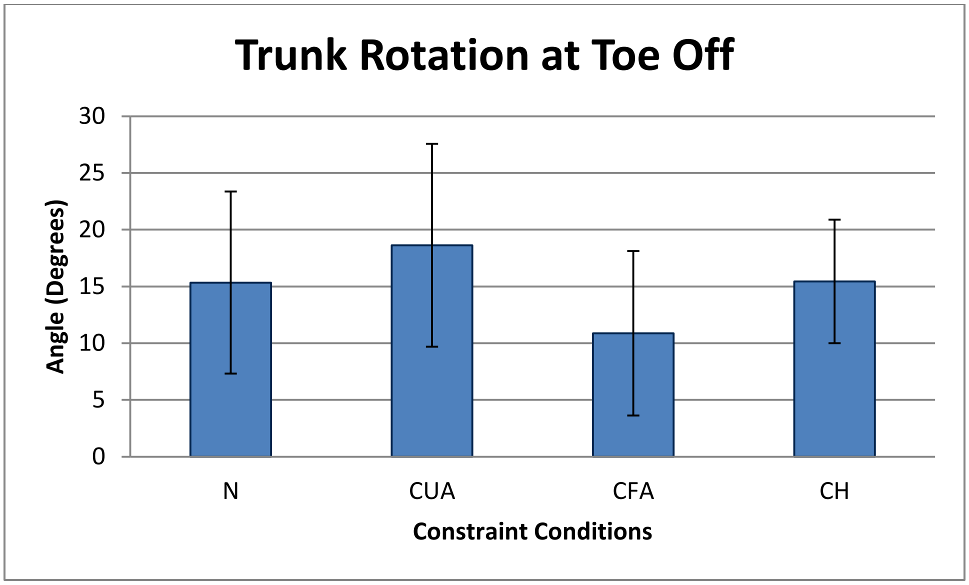

Figure 37. Effects of constraint on trunk angle in the transverse plane at toe off.

While continuing to move superiorly in the kinetic chain, the upper extremity joints were statistically analyzed to determine the effects of constraint at the IL shoulder and elbow joints at initial contact and toe off. A 1 (shoulder angle, SA_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$ to determine the influence of constraint on shoulder angle at initial contact. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of shoulder angle (SA_{IC}) ($F(1,14) = 3.147$, $p = 0.098$, $\eta^2 = 0.184$, *Power* = 0.379). Shoulder data at initial contact were represented as angle of the upper arm with reference to the trunk with positive values indicating shoulder extension. A position of extension was noted at the shoulder for all the participants at initial contact. Although there was no significant difference between the four constraint conditions, the CH condition yielded a larger SA_{IC} indicating greater shoulder extension at initial contact during the non-constrained condition compared to N, CUA and CFA (Figure 38).

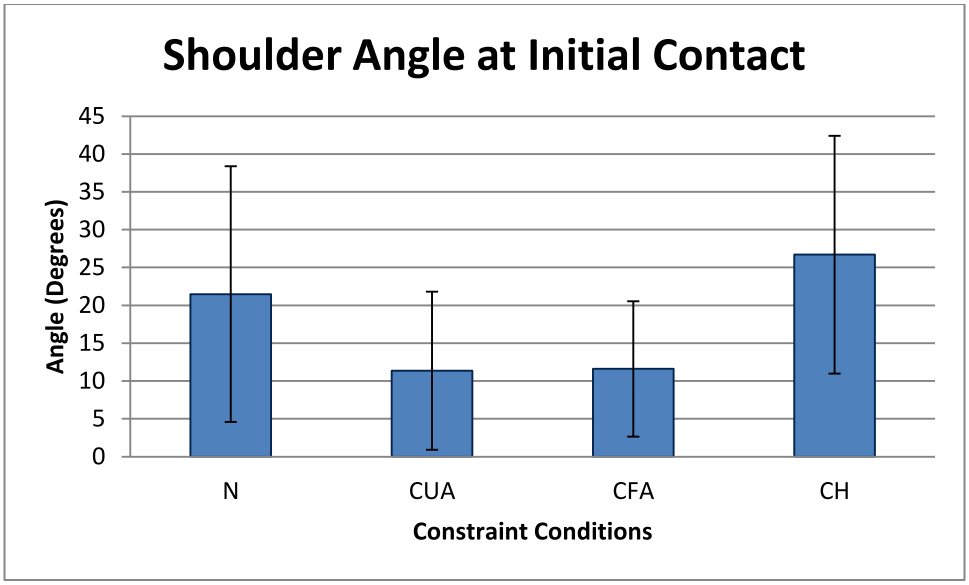

Figure 38. Effects of constraint on shoulder angle at initial contact.

In order to determine the influence of constraint on shoulder angle at toe off, a 1 (shoulder angle, SA_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of shoulder angle (SA_{TO}) ($F(1,14) = 15.977$, $p = 0.001$, $\eta^2 = 0.533$, *Power* = .960). This indicates that different constraint conditions altered shoulder angle at toe off. Follow-up pairwise comparisons for SA_{IC} were conducted and significant differences were noted between the following conditions: N and CUA (p = .001), N and CFA (p = .000), CUA and CFA (p = .000), CUA and CH (p = .007), and CFA and CH (p = .000) (Figure 39). Shoulder data at toe off are represented as degrees of flexion with positive values indicated shoulder flexion. However, during the CFA condition, the shoulder experienced extension which is represented in negative degrees of flexion. Results from the analysis indicate that the shoulder was in extension at toe off during toe off in the CFA condition. N, CUA, and CH demonstrated significantly greater shoulder flexion at toe off with N producing the greatest amount of flexion, followed by CH and CUA.

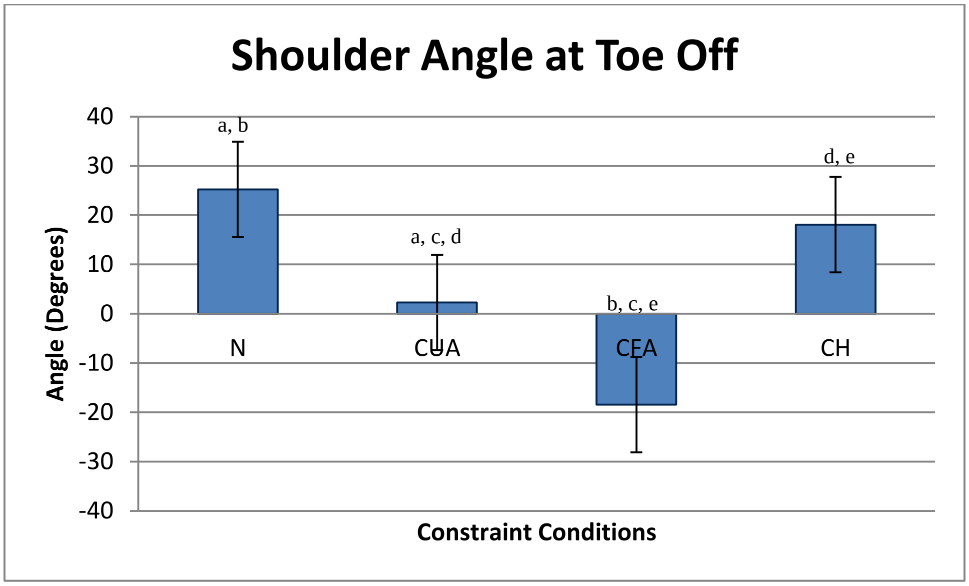

Figure 39. Effects of constraint on shoulder angle at toe off.
N-CUA, N-CFA, CUA-CFA, CUA-CH, CFA-CH: [a,b,c,d,e] p<.05.

In addition, in order to determine the effects of constraint on the elbow angle at initial contact, a 1 (elbow angle, EA_{IC}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position elbow angle (EA_{IC}) ($F(1,14) = 2.798$, $p = 0.117$, $\eta^2 = 0.167$, *Power* = 0.344). Elbow data at initial contact are represented as the degrees between the forearm and upper arm. Although there was no significant difference between the four constraint conditions, the CFA condition yielded the greatest EA_{IC} indicating greater elbow extension at initial contact during the constrained condition in comparison to N, CUA and CH (Figure 40).

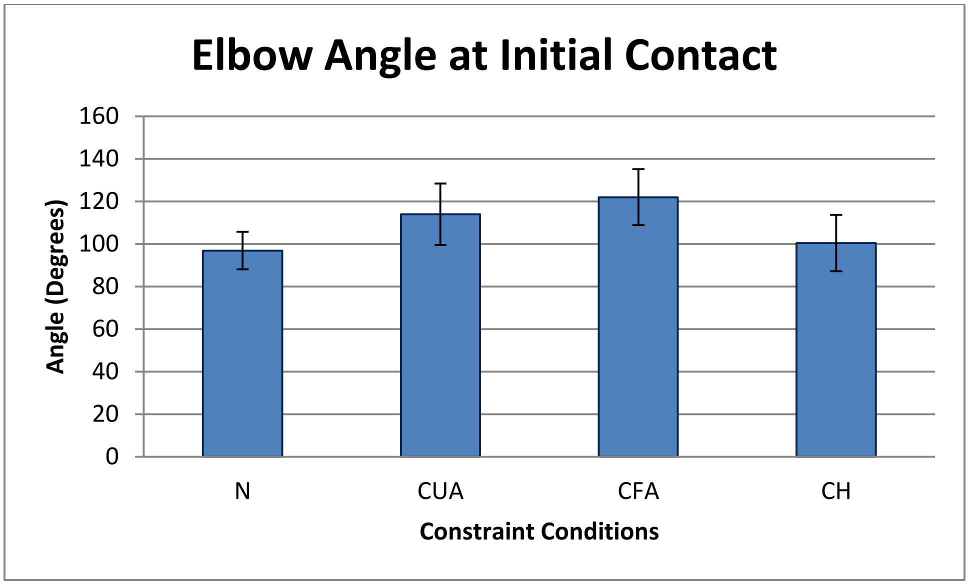

Figure 40. Effects of constraint on elbow angle at initial contact.

Furthermore, to determine the influence of constraint on elbow angle at toe off, a 1 (elbow angle, EA_{TO}) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the joint position of elbow angle (EA_{TO}) ($F(1,14) = 97.123$, $p = 0.000$, $\eta^2 = 0.874$, *Power* = 1.00). Elbow data at initial contact are represented as the degrees between the forearm and upper arm. This indicates that different constraint conditions altered elbow angle at toe off. Follow-up pairwise comparisons for EA_{TO} were conducted and significant differences were noted between the following conditions: N and CUA (p = .018), N and CFA (p = .000), CUA and CFA (p = .000), and CFA and CH (p = .000) (Figure 41). These results indicate that the IL elbow had significantly greater extension at toe off in the CFA condition when compared to N, CUA and CH. Moreover, N demonstrated the greatest amount of flexion followed by CH and CUA.

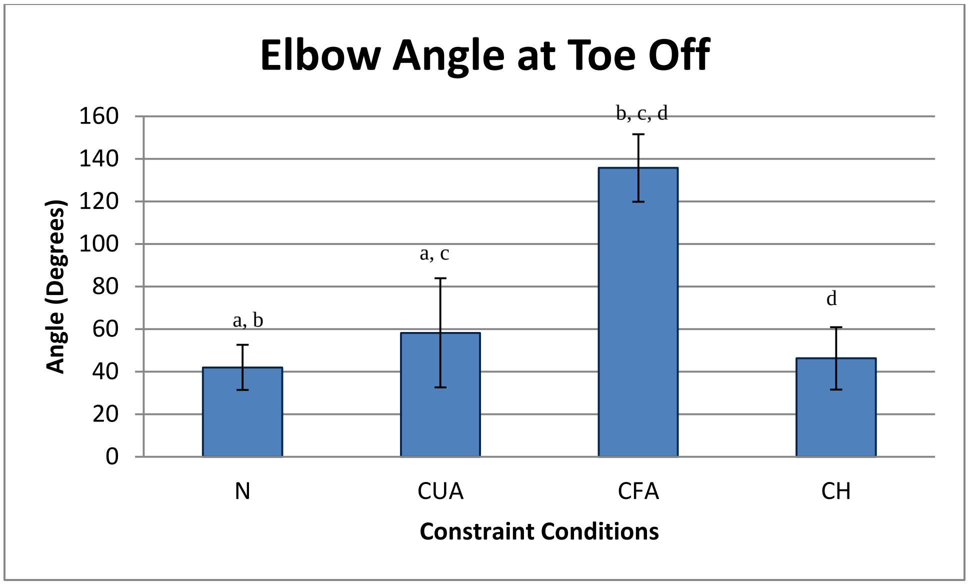

Figure 41. Effects of constraint on elbow angle at toe off.
N-CUA, N-CFA, CUA-CFA, CFA-CH: ^{a,b,c,d} p<.05.

Section 4: Ground Reaction Forces and Impulses

In addition to examining the effects of constraint conditions CUA, CFA and CH on spatial-temporal and joint position, ground reaction forces and impulses were also investigated during right foot contact. This section includes the results from the N, CUA, CFA and CH conditions on $VGRF_{PEAK}$, $VGRF_{TTP}$, VGRI, $BGRF_{PEAK}$, BGRI, $PGRF_{PEAK}$, and PGRI, including follow-up tests for significant differences between conditions.

In order to determine the influence of constraint on peak vertical ground reaction force, a 1 (force, $VGRF_{PEAK}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on peak vertical ground reaction force ($VGRF_{PEAK}$) ($F(1,14) = 1.057$, $p = 0.321$, $\eta^2 = 0.070$, *Power* = 0.160). Although there was no significant difference between the four constraint conditions, the N condition yielded a larger $VGRF_{PEAK}$ which is indicative of greater vertical ground reaction force when compared to CUA, CFA, and CH conditions (Figure 42).

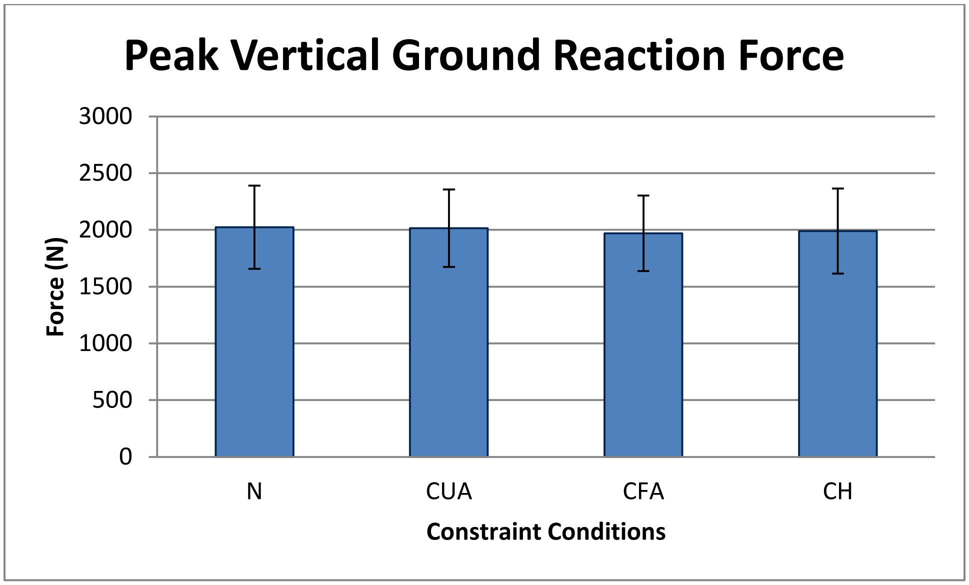

Figure 42. Effects of constraint on peak vertical ground reaction force.

In addition, a 1 (time, $VGRF_{TTP}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$ to determine the influence of constraint on time to peak vertical ground reaction force. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on time to peak vertical ground reaction force ($VGRF_{TTP}$) ($F(1,14) = 2.36$, $p = 0.634$, $\eta^2 = 0.017$, *Power* = 0.074). Although there was no significant difference between the four constraint conditions, the CH condition yielded a greater $VGRF_{TTP}$ indicating peak vertical ground reaction force was reached sooner during the constrained hip condition (Figure 43).

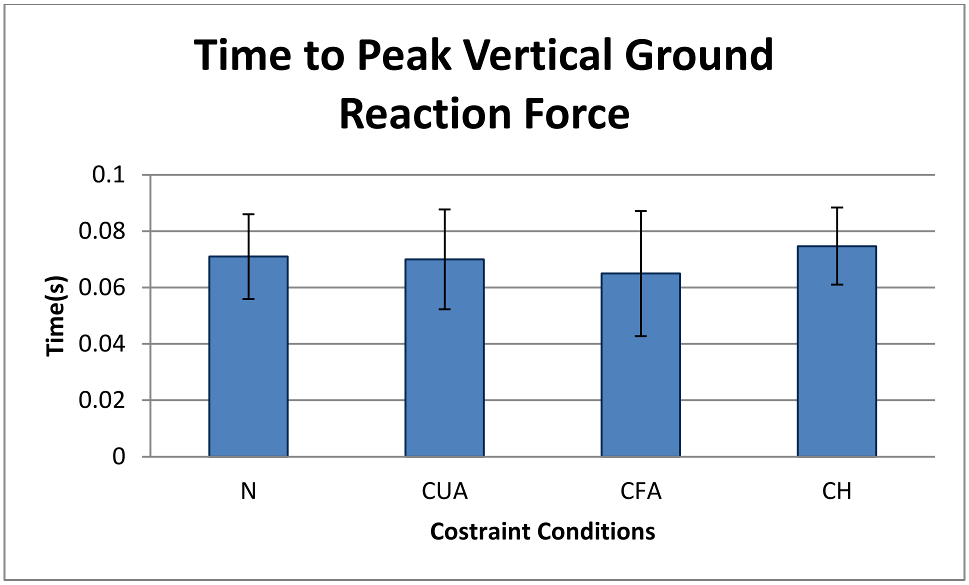

Figure 43. Effects of constraint on time to peak vertical ground reaction force.

Furthermore, to determine the effects of constraint on vertical ground reaction impulse, a 1 (impulse, VGRI) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on vertical ground reaction impulse (VGRI) ($F(1,14) = .591$, $p = 0.455$, $\eta^2 = 0.041$, *Power* = 0.111). Although there was no significant difference between the four constraint conditions, the CUA condition yielded the largest VGRI indicating greater impulse during the condition (Figure 44).

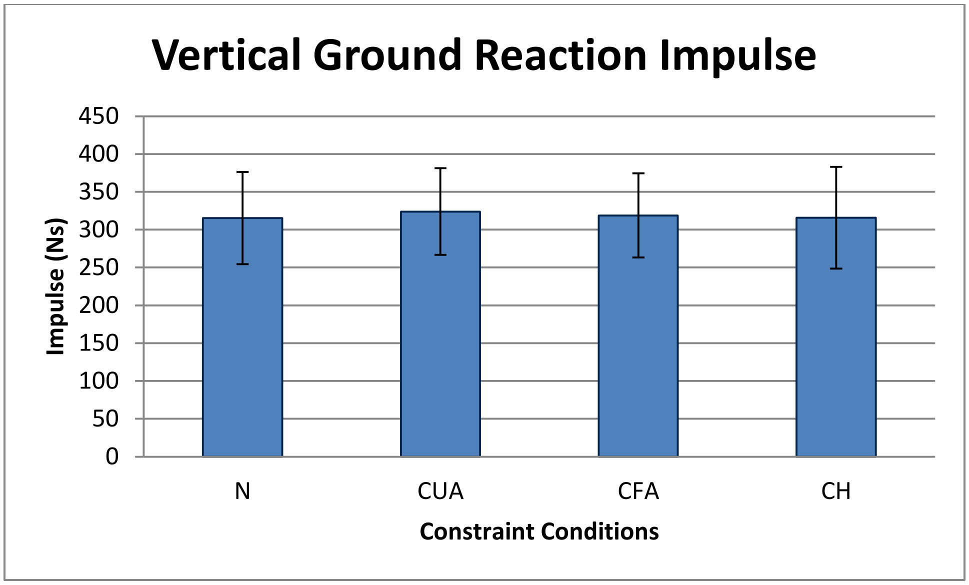

Figure 44. Effects of constraint on vertical ground reaction impulse.

In addition to ground reaction force and impulse in the vertical direction, braking a propulsive forces and impulses were analyzed. In order to determine the effects of constraint on peak braking ground reaction force, a 1 (force, $BGRF_{PEAK}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on peak braking ground reaction force ($BGRF_{PEAK}$) ($F(1,14) = 18.225$, $p = 0.001$, $\eta^2 = 0.566$, *Power* = 0.977). This indicates that different constraint conditions altered peak braking ground reaction force. Follow-up pairwise comparisons for $BGRF_{PEAK}$ were conducted and significant differences were noted between the following conditions: N and CH (p = .001), and CUA and CH (p = .001) (Figure 45). CH demonstrated significantly lower peak braking ground reaction force with 312.99N of force, in comparison to N and CUA. $BGRF_{PEAK}$ was the greatest in N, followed by CUA then CFA.

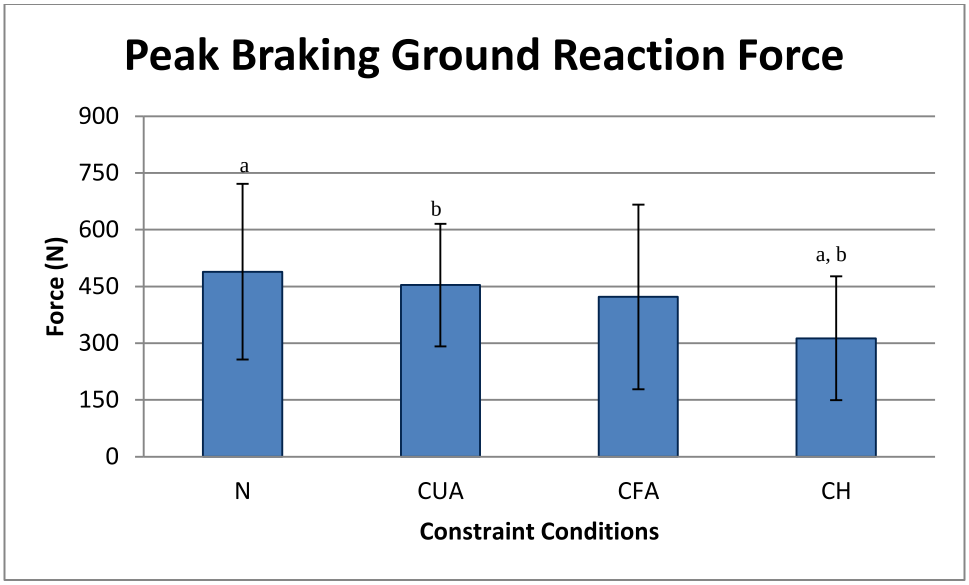

Figure 45. Effects of constraint on peak braking ground reaction force.
N-CH, CUA-CH: [a,b] p<.05.

In order to determine the influence of constraint on braking ground reaction impulse, a 1 (impulse, BGRI) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a p value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on braking ground reaction impulse (BGRI) ($F(1,14) = 5.852$, $p = 0.030$, $\eta^2 = 0.295$, *Power* = 0.615). This indicates that different constraint conditions altered braking ground reaction impulse. Follow-up pairwise comparisons for BGRI were conducted and significant differences were noted between the following conditions: N and CH (p = .028), CUA and CH (p = .022), and CFA and CH (p = .029) (Figure 46). BGRI was significantly lowest in CH and N demonstrated the greatest BGRI, followed by CUA and CFA.

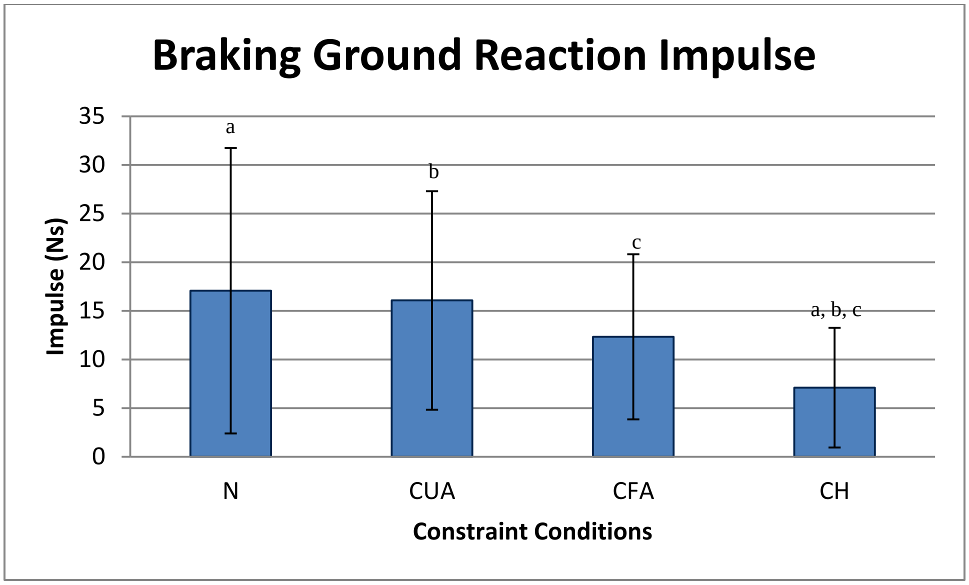

Figure 46. Effects of constraint on braking ground reaction impulse.
N-CH, CUA-CH, CFA-CH: ^{a,b,c} $p<.05$.

Furthermore, in order to determine the influence of constraint on peak propulsive ground reaction force, a 1 (force, $PGRF_{PEAK}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on peak propulsive ground reaction force ($PGRF_{PEAK}$) ($F(1,14) = .718$, $p = 0.411$, $\eta^2 = 0.049$, *Power* = 0.124). Although there was no significant difference between the four constraint conditions, the CH condition yielded a larger $PGRF_{PEAK}$ indicating greater propulsive ground reaction force during the constrained condition (Figure 47).

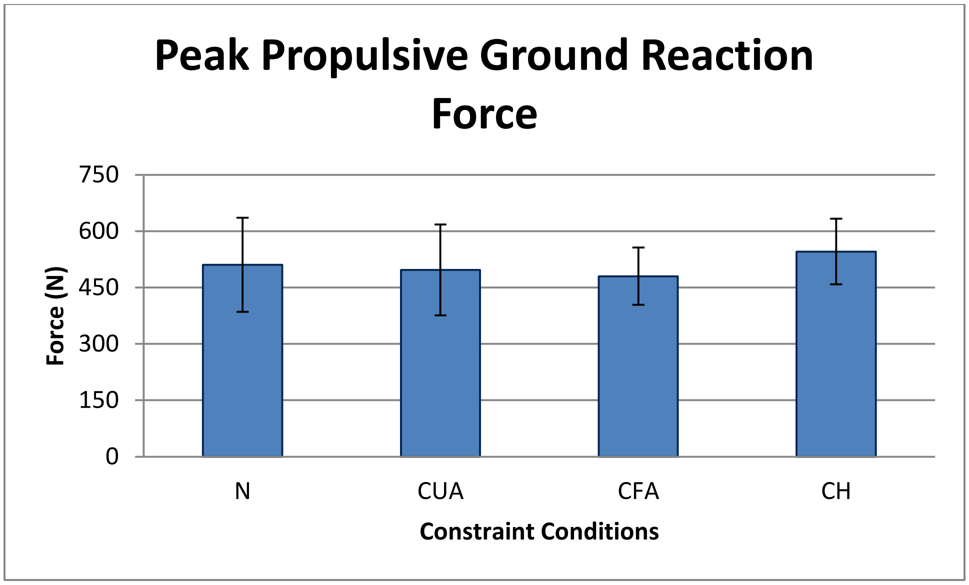

Figure 47. Effects of constraint on peak propulsive ground reaction force.

In addition, a 1 (impulse, PGRI) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$ to determine the effects of constraint on propulsive ground reaction impulse. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on propulsive ground reaction impulse (PGRI) ($F(1,14) = 2.153$, $p = 0.164$, $\eta^2 = 0.133$, *Power* = 0.277). Although there was no significant difference between the four constraint conditions, the CH condition yielded a larger PGRI indicating greater propulsive ground reaction impulse during the hip constrained condition (Figure 48).

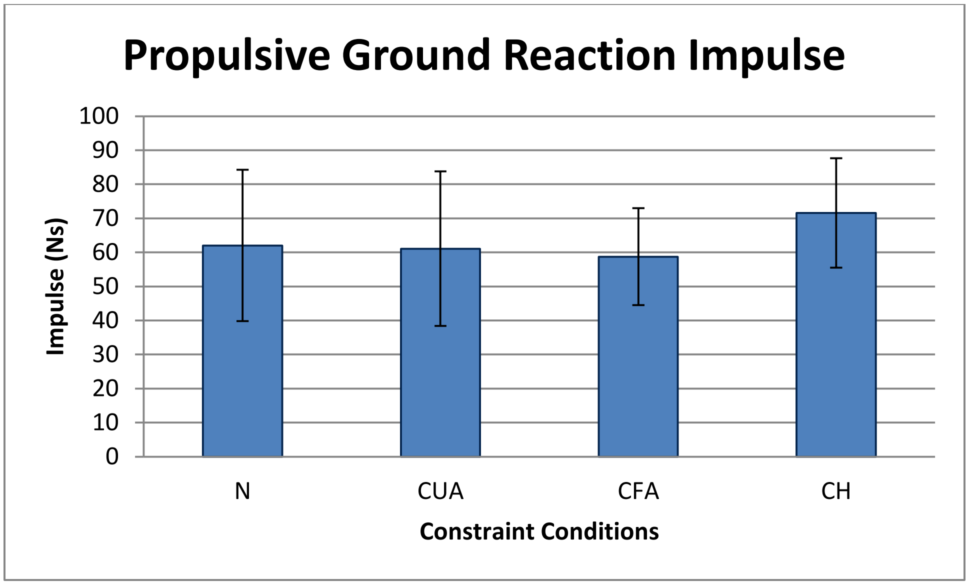

Figure 48. Effects of constraint on propulsive ground reaction impulse.

Section 5: Electromyography

In addition to examining the effects of constraint conditions CUA, CFA and CH on spatial-temporal and joint position and gait kinetics, muscle activity at the IL LD and GM and CL LD and GM were investigated at initial contact and toe off of the right foot. This section includes the results from the N, CUA, CFA and CH conditions on peak activity of the IL LD at initial contact and toe off, peak activity of the CL LD at initial contact and toe off, peak activity of the IL GM at initial contact and toe off and peak activity of the CL GM at initial contact at toe off ($ILLD_{IC}$, $ILLD_{TO}$, $CLLD_{IC}$, $CLLD_{TO}$, $ILGM_{IC}$, $ILGM_{TO}$, $CLGM_{IC}$ and $CLGM_{TO}$, respectively) including follow-up tests for significant differences between conditions. All sEMG data are represented as peak values as a percentage of the MVIC of the participant.

To investigate the influence of constraint on the peak muscle activity of the IL LD at initial contact, a 1 (muscle activity, $ILLD_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the IL LD at initial contact ($ILLD_{IC}$) ($F(1,14) = 2.6$, $p = 0.129$, $\eta^2 = 0.157$, *Power* = 0.324). Although there was no significant difference between the four constraint conditions, the CH condition yielded the largest $ILLD_{IC}$ activity indicating greater IL LD activity at initial contact of the right foot during the hip constrained condition. In addition, IL LD muscle activity was the lowest during the N condition (Figure 49).

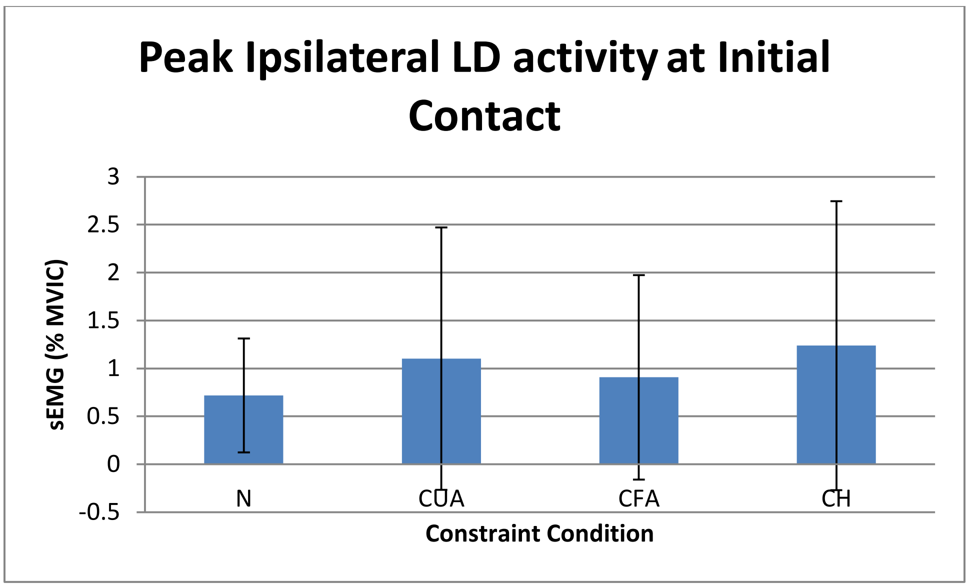

Figure 49. Effects of constraint on peak IL LD activity at initial contact.

To investigate the influence of constraint on the peak muscle activity of the IL LD at toe off, a 1 (muscle activity, $ILLD_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a p value set apriori at $p < 0.05$. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the IL LD activity ($ILLD_{TO}$) ($F(1,14) = 5.37$, $p = 0.36$, $\eta^2 = 0.277$, *Power* = 0.578). This indicates that different constraint conditions altered muscle activity of the IL LD at toe off. Follow-up pairwise comparisons for $ILLD_{TO}$ were conducted and significant differences were noted between the following conditions: N and CH (p = .029), CUA and CH (p = .007), and CFA and CH (p = .002) (Figure 50). These results indicate that muscle activity of the IL LD at toe off was significantly greater in the CH condition when compared to the N, CUA and CFA conditions. Interestingly, CFA indicated the lowest peak muscle activity of the IL LD at toe off.

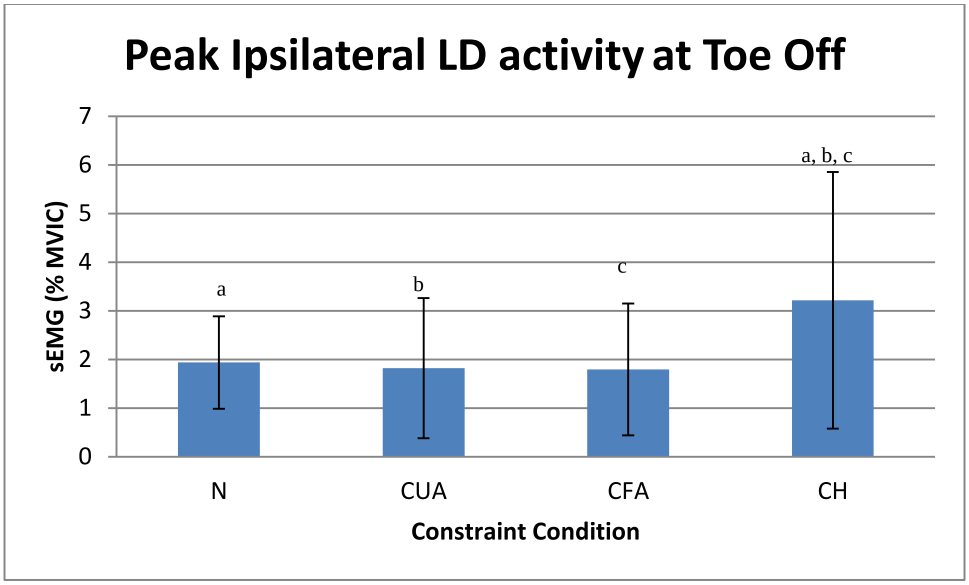

Figure 50. Effects of constraint on peak IL LD activity at toe off.
N-CH, CUA-CH, CFA-CH: [a,b,c] p<.05

In addition to the IL LD, the CL muscle was examined. To investigate the influence of constraint on the peak muscle activity of the CL LD at initial contact, a 1 (muscle activity, $CLLD_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the CL LD at initial contact ($CLLD_{IC}$) ($F(1,14) = 0.355$, $p = 0.561$, $\eta^2 = 0.025$, *Power* = 0.086). Although there was no significant difference between the four constraint conditions, the CH condition yielded a greater $CLLD_{IC}$ indicating higher peak muscle activity during the hip constrained condition. In addition, muscle activity for condition CFA was the lowest across all conditions (Figure 51).

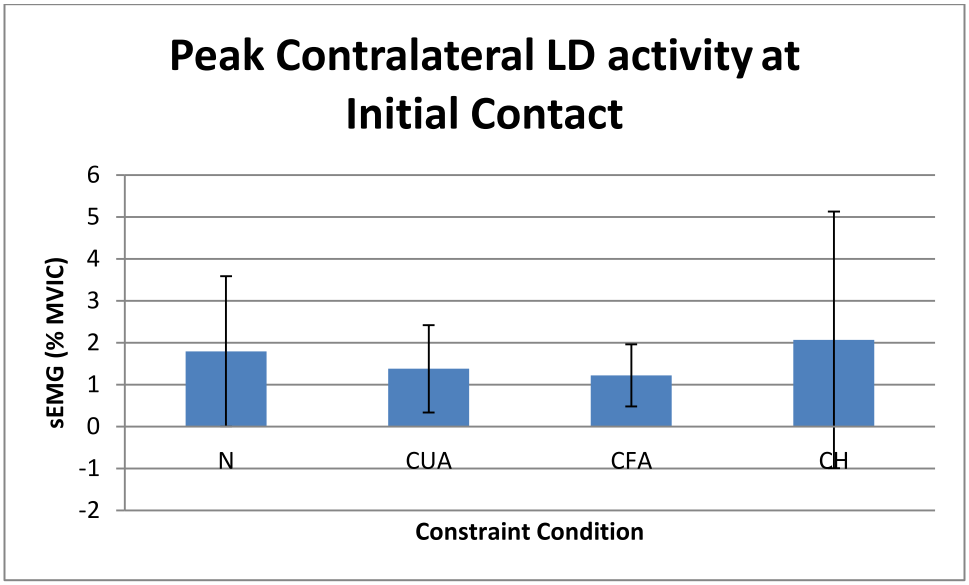

Figure 51. Effects of constraint on peak CL LD activity at initial contact.

To investigate the influence of constraint on the peak muscle activity of the CL LD at toe off, a 1 (muscle activity, $CLLD_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p <0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the CL LD at toe off ($CLLD_{TO}$) ($F(1,14) = .712$, $p = 0.413$, $\eta^2 = 0.048$, *Power* = 0.123). Although there was no significant difference between the four constraint conditions, the CH condition yielded a larger $CLLD_{TO}$ indicating greater activity at toe off during the hip constrained condition. In addition, muscle activity for condition CFA was the lowest (Figure 52).

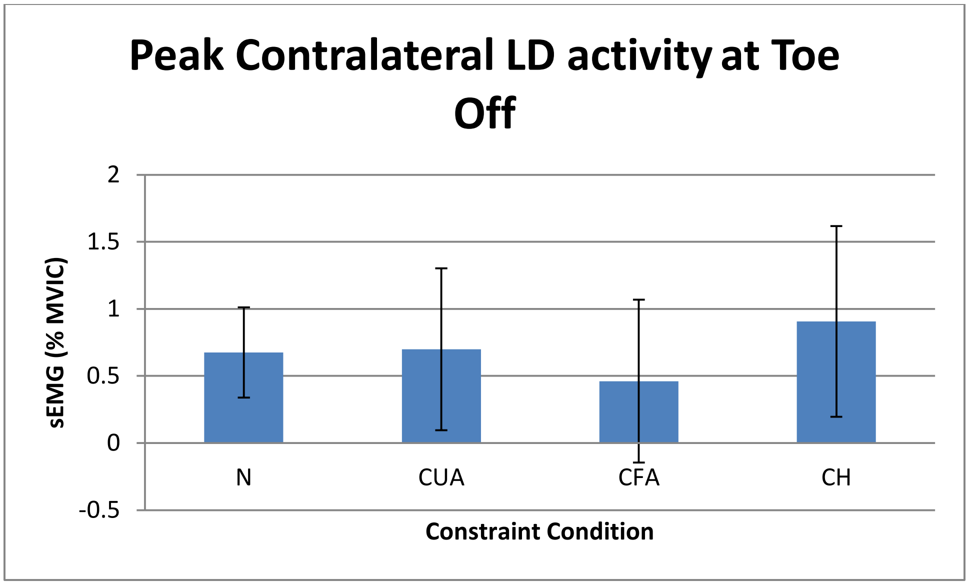

Figure 52. Effects of constraint on peak CL LD activity at toe off.

In addition to the bilateral LD activities, bilateral GM muscle activity was also recorded. To investigate the influence of constraint on the peak muscle activity of the IL GM at initial contact, a 1 (muscle activity, $ILGM_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the IL GM at initial contact ($ILGM_{IC}$) ($F(1,14) = 2.276$, $p = 0.154$, $\eta^2 = 0.14$, *Power* = 0.29). Although there was no significant difference between the four constraint conditions, the CH condition yielded the lowest peak GM activity (Figure 53).

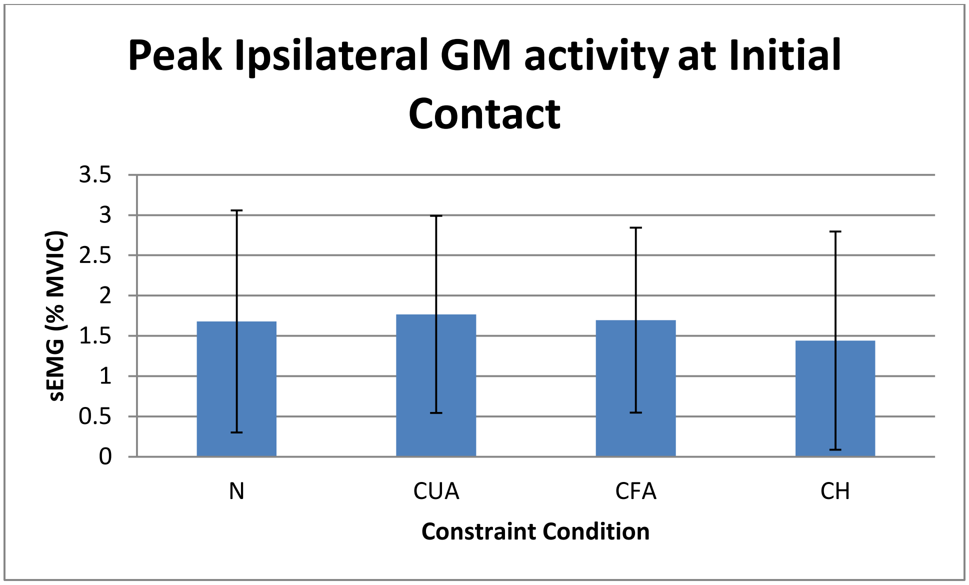

Figure 53. Effects of constraint on peak IL GM activity at initial contact.

To investigate the influence of constraint on the peak muscle activity of the IL GM at toe off, a 1 (muscle activity, $ILGM_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the IL GM at toe off ($ILGM_{TO}$) ($F(1,14) = 0.111$, $p = 0.744$, $\eta^2 = 0.008$, $Power = 0.061$). Although there was no significant difference between the four constraint conditions, the CH condition yielded the lowest peak activity of the GM, similarly to initial contact. In addition, muscle activity of the IL GM for condition CUA was the highest across all conditions (Figure 54).

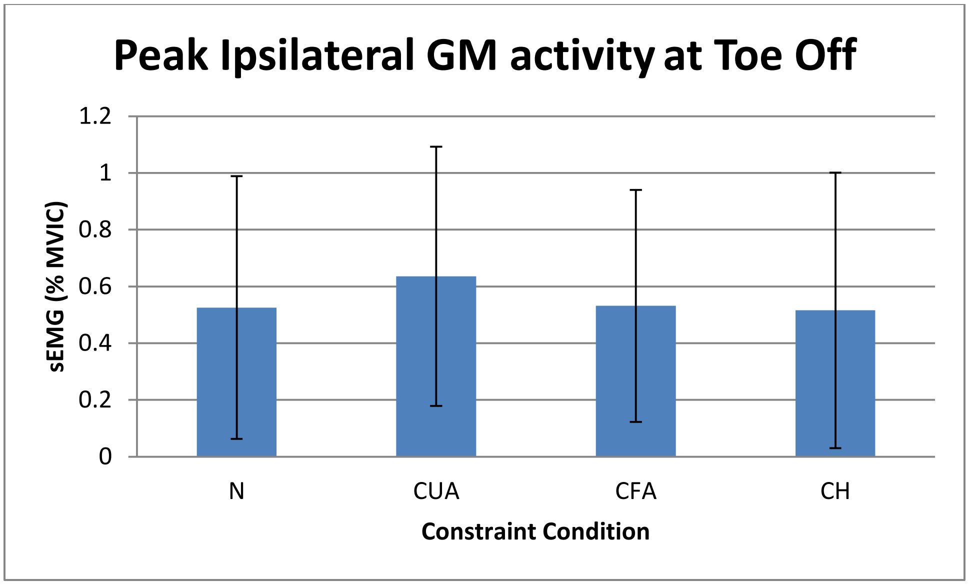

Figure 54. Effects of constraint on peak IL GM activity at toe off.

In addition to the IL GM, the CL GM was analyzed. To investigate the influence of constraint condition on the peak muscle activity of the CL GM at initial contact, a 1 (muscle activity, $CLGM_{IC}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at $p < 0.05$. The results yielded a non-significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the CL GM at initial contact ($CLGM_{IC}$) ($F(1,14) = 0.491$, $p = 0.495$, $\eta^2 = 0.034$, *Power* = 0.1). Although there was no significant difference between the four constraint conditions, the CH condition yielded lower peak activity of the GM at toe off (Figure 55).

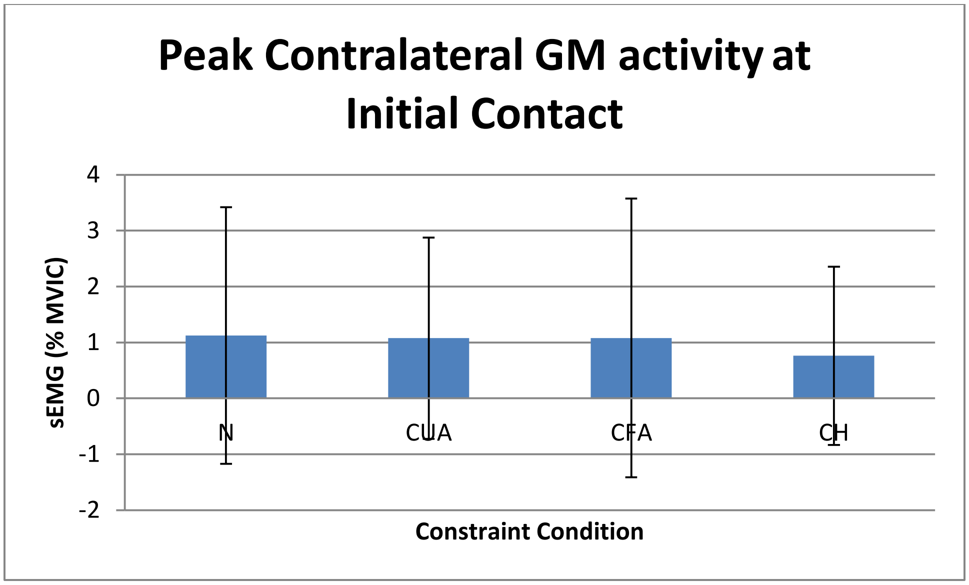

Figure 55. Effects of constraint on peak CL GM activity at initial contact.

To investigate the influence of constraint condition on the peak muscle activity of the CL GM at toe off, a 1 (muscle activity, $CLGM_{TO}$) x 4 (constraint condition, N, CUA, CFA, CH) repeated measures ANOVA was conducted with a *p* value set apriori at p <0.05. The results yielded a significant main effect for constraint condition (N, CUA, CFA and CH) on the muscle activity of the CL GM at toe off ($CLGM_{TO}$) ($F(1,14) = 7.57$, $p = 0.016$, $\eta^2 = 0.351$, *Power* = 0.725). This indicates that different constraint conditions altered muscle activity of the CL GM at toe off. Follow-up pairwise comparisons for $CLGM_{TO}$ were conducted and significant differences were noted between conditions N and CH (p = .011) (Figure 56). These results indicate that peak muscle activity of the GM is significantly lower in the CH condition when compared to the N condition

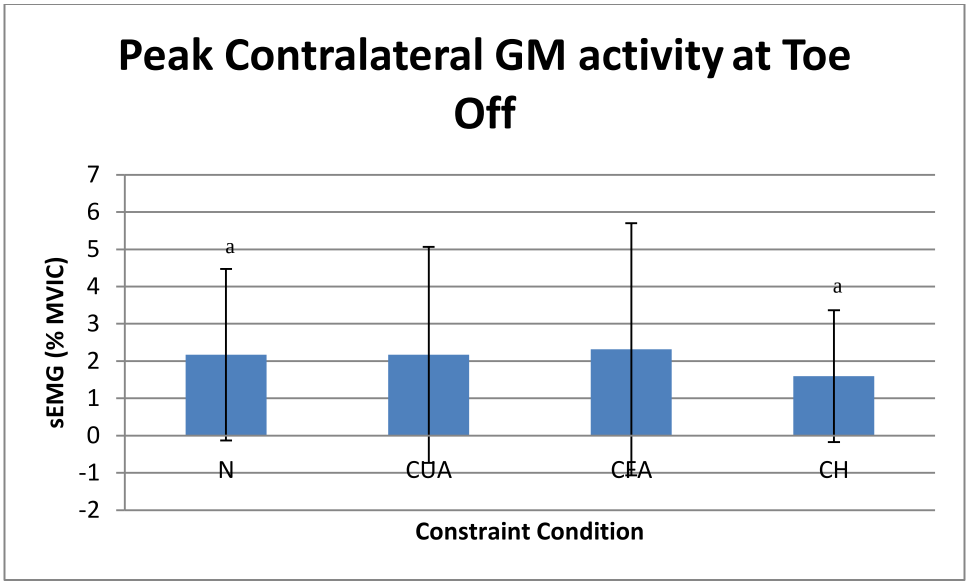

Figure 56. Effects of constraint on peak CL GM activity at toe off.
N-CH: [a] p<.05.

CHAPTER V

DISCUSSION

The purpose of this investigation was two-fold: 1) to determine the role of the lumbopelvic-hip complex (LPHC) during the initial acceleration of a sprint and 2) to determine the overall effects of different arm and leg constraints on sprinting mechanics by examining the spatial-temporal and joint kinematics, kinetics and muscular activity, specifically the latissimus dorsi (LD) and gluteus maximus (GM), at different points during bipedal locomotion acceleration.

The following chapter has been divided into five primary sections: Section 1: The effect constrained upper arm (CUA), constricted full arm (CFA) and constricted hip (CH) conditions versus the normal (N) condition on spatial-temporal kinematics, Section 2: The effect of the CUA, CFA and CH conditions on joint kinematics compared to normal (N) condition, Section 3: The effect of the CUA, CFA and CH conditions on ground reaction forces and impulses compared to normal (N) condition, and Section 4: The effect of the CUA, CFA and CH conditions on LD and GM activity compared to normal (N) condition. The final section includes conclusions and suggestions for future research.

Section 1: N, CUA, CFA and CH Conditions on Spatial-Temporal Kinematics

This section will discuss the results regarding the influence of constraint conditions (CUA, CFA and CH) on the spatial-temporal kinematics, specifically, step lengths (SL_R, SL_L), contact times (CT_R, CT_L), and velocity (VEL) of the participants when compared to a normal, non-constrained condition (N). Figure 57 illustrates constraint conditions compared to normal for each kinematic variable measure.

Figure 57. *Spatial-Temporal Kinematics*

Kinematic Variable	CUA	CFA	CH
SL_R	-	-	-
SL_L	-*	-*	-*
NST_R	-	+	-
NST_L	+	+	-
CT_R	+	+	+
CT_L	+*	+*	+*
VEL	+*	+*	+*

Note. + = Kinematic variable larger than N condition (larger step length, larger contact time, and larger velocity). - = Kinematic variable smaller than N condition (smaller step length, smaller contact time, and lower velocity). * = Significant difference from N condition.

Step Lengths

Stride length is a commonly studied component of gait. However in running, step length is more frequently considered as stride lengths may exceed capture volumes and encapsulate only one flight phase. Several studies have identified changes in step length during the acceleration phase of a sprint (Coh et al., 2006; Kyrolainen et al., 1999). Specifically, Coh and colleagues (2006) observed step length progressively increase from step 1 to step 10 from 103 cm, to 186 cm, respectively. In addition, Kyrolainen et al

(1999) also noted step length decreases as running speed increased further supporting step lengths increase during acceleration.

During constant velocity running, step length was affected by arm constraints. In particular, Frere et al. (2009) found that when the arms are constrained from typical motion, step length decreased. Likewise, Sayers (1998) found that when the arms were constrained step length also decreased. While these studies provided valuable insight into influences on step length, the current study is a novel investigation to evaluate the influence of acceleration and arm/hip constraint on step length.

In the present study there was no significant difference found between conditions on the fourth step of the acceleration, on the right foot (SL_R). However, step lengths in the constraint condition were smaller when compared to the N condition. In addition, step length 5 was greater than step length 4 within each condition, supporting the findings of Coh et al. (2006) and Kyrolainen et al. (1999). The findings of the present study support the progressive increase in step length during acceleration, as noted by the increase between SL_R and SL_L, which represented steps 4 and 5, respectively. While it is surprising that there was no significance observed for the 4th step, a significant difference was noted in the 5th step. It is possible that the 5th step represents a transition point beyond which the differences between the constrained and non-constrained step lengths would get progressively significantly different and subsequent steps would increase in length (up to constant velocity running).

There was a significant main effect for constraint condition on the 5th step of the bout that occurred on the left foot, SL_L. Left step length was significantly shorter in all the constraint conditions compared to the normal (Figure 57), with the shortest step length occurring during CH (Figure 14). The shorter SL_L suggests that CH decreases step lengths to a greater extent than with CUA and CFA, but each constraint condition decreased step length. Mean lengths for SL_L for all conditions were the following: N (1.61m), CUA (1.54m), CFA (1.49m), and CH (1.48m). As both step length and elbow angle were significantly different between CUA and CFA, it can be concluded that elbow motion contributed to the longer step length.

Non-Support Times

While an athlete is sprinting, there is a period in the gait cycle where neither foot is in contact with the ground. Previous research indicates that this non-support phase increases in duration as horizontal velocity increases (Atwater, 1982; Coh et al., 2006; Hunter et al., 2004). Specifically, Coh et al. (2006) discovered average flight time at step 1 from a block start was 50 ms, while step 10 demonstrated a period of 115. In addition, Hunter et al. (2004) found that flight time increased 102 – 121 ms from the start to 16 m.

There were no significant differences found between conditions of non-support time prior to right foot contact (NS_R) (Figure 15) or non-support time prior to left foot contact (NS_L) (Figure 16). CH demonstrated the shortest period of non-support prior to foot contact, while CFA generated the longest period of non-support. This suggests that since the hips were limited in range of motion, the foot did not travel as long in the air as it did during the 3 other conditions. CH continued to have the shortest period of non-support prior to left foot contact, suggesting that the lack of hip motion decreases the period of non-support, thus inhibiting the ability of the athlete to drastically increase the non-support time. There was nearly no change between the arm constraint conditions and the N, suggesting that the arm motion does not dramatically affect non-support time.

Contact Times

Contact time is defined as the period where either of the feet is in contact with the ground (Johnson & Buckley, 2001) and has been well versed in literature investigating bipedal acceleration stating that a progressive decrease in contact time is beneficial to acceleration (Atwater, 1982; Coh et al., 2006; Moravec et al., 1988). For example, Moravec and colleagues (1988) stated as velocity increases, contact time decreases. In addition, Atwater (1982) observed that contact times at first and second ground contacts are correlated with mean horizontal velocity during the initial acceleration phase of a sprint ($r = -0.65$ and $r = -0.44$, respectively). Specifically, contact times ranged from 170 - 230 ms for the first ground contact and 150 – 190 ms for the second ground contact after the start of a block sprint. Furthermore, Coh and colleagues (2006) observed contact time at step 1 to be 177 ms, while contact time at step 10 was 110 ms, representing a progressive decrease during the acceleration of a sprint. On the other hand, Miller et al., 2009 found that the implementation of arm constraints yielded an increase in the duration of contact phase. The present investigation attempted to further elucidate these findings by combining a condition in which contact time should decrease (early acceleration) and a condition in which contact time should increase (arm constraint).

Results from the current study follow the trend from literature indicating contact times progressively decrease during acceleration (Figure 17 and Figure 18). In addition, contact time of the left foot (CT_L) yielded a significant main effect between all conditions and N. N presented the shortest contact time and CFA presented the longest time (Figure 57). Mean times for CT_L for all conditions were as follows: N (.14s), CUA (.15s), CFA (.16s), and CH (.16s). The findings suggest that while the entire arm was constrained, contact time was the greatest meaning the foot was in contact with the ground for the longest amount of time, synonymous with literature (Miller et al., 2009). The CFA condition yielded a longer contact time than did the CUA condition suggesting that elbow motion influenced contact time. This is further supported by a larger step length and velocity noted during the CUA condition. Last, the CH condition also yielded a significantly longer contact time when compared to the N condition, suggesting that the decrease in hip range of motion requires the foot to be in contact with the ground longer. In light of the velocity data presented in the next section, it would appear that the longer contact time is associated with smaller velocity.

Velocity

Literature has found that velocity of the runner increases during this early acceleration phase of a sprint (Atwater, 1982; Coh et al., 2006; Hunter et al., 2004; Kyrolainen et al., 1999; Mann & Herman, 1985; Mero et al., 1992; Moravec et al., 1988; Murphy et al., 2003). However, a decrease in velocity has been observed while examining the effects of arm constraint on locomotion (Frere et al., 2009; Grant et al., 2003; Miller et al., 2009; Widowski & Gittoes, 2012). By investigating the acceleration phase of sprinting with the inclusion of arm/hip constraints the present study endeavored to identify the influence of conditions known to diminish velocity (constraints) during an activity marked by increases in velocity.

In the present study, a significant main effect of constraint condition on velocity (VEL) was observed (Figure 19). VEL was significantly greater in condition N when compared to all constraint groups (Figure 57). Mean VEL for each condition was as follows: N (6.1 m/s), CUA (5.89 m/s), CFA (5.71 m/s) and CH (5.68 m/s). These findings support the literature that has identified a decrease in velocity with arm constraints and expands the literature by identifying hip constraint as a detriment to an increase in velocity. Furthermore, as the level of arm constraint increased, the VEL decreased, suggesting that arm swing plays a considerable role in velocity development.

Summary

Results from the current study follow the trend of literature that has investigated spatial-temporal measures of acceleration locomotion and arm constraint research on gait kinematics. However, to the author's knowledge, this was the first study that has incorporated different constraint conditions in an effort to measure the effects on spatial-temporal kinematics during bipedal locomotion acceleration. It was noted that significant differences between constraint conditions and non-constraint were found for SL_L, CT_L and VEL. Decreases in step length and velocity were not surprising for the CH condition, however the differences noted during the arm constraint conditions reinforces the role of arm swing on speed as well as the components that comprise speed (step length and contact time). SL_L significantly decreased in all constraint conditions when compared to the normal, unconstrained condition, suggesting that the decrease that was seen in SL_L also decreased VEL in the constraint conditions. In addition, CT_L increased in constraint conditions which further complete the connection to the significant decrease in VEL noted with the constraint conditions. In essence, a greater SL_L and smaller CT_L yielded the greatest VEL, which was observed in the N condition. Similarly, the CFA and CH conditions yielded the two smallest SL_L and CT_L, and presented the lowest VEL between all 4 conditions. Typically, as the period of non-support increases, the duration of contact time decreases, however, the findings of the present study suggest that the CFA condition increased non-support time yet increased contact time as well. Additional pairwise comparisons between CUA and CFA presented a significant decrease in VEL in the CFA condition. It is suggested that the lack of elbow motion in CFA contributed to a lower velocity. Further investigation is necessary to help understand the connection seen with this constraint and respective variables. Furthermore, the negative influence of the CH condition on VEL indicates that hip range of motion is important in the development of velocity.

Section 2: N, CUA, CFA and CH Conditions on Joint Kinematics

This section will discuss the results regarding the influence of constraint conditions CUA, CFA and CH on joint kinematics at the ipsilateral (IL) ankle (AA_{IC}, AA_{TO}), IL knee (KA_{IC}, KA_{TO}), IL hip (HA_{IC}, HA_{TO}), pelvis ($APPGR_{IC}$, $APPGR_{RO}$, $TVPGR_{IC}$, $TVPGR_{TO}$, $LPGR_{IC}$, $LPGR_{TO}$), trunk (TF_{IC}, TF_{TO}, LTF_{IC}, LTF_{TO}, TR_{IC}, TR_{TO}), IL shoulder (SA_{IC}, SA_{TO}) and IL elbow (EA_{IC}, EA_{TO}) during initial contact and toe off of the fifth step of the acceleration, which occurred on the right foot, in comparison to the N condition. This section is divided between lower extremity joints, pelvic girdle, trunk, and upper extremity joints. The discussion will be ordered such that all joint position data at initial contact (Figure 58) will be discussed first followed by joint position data at toe off (Figure 59). Figures 60 and 61 illustrate ankle, knee, hip, shoulder and elbow angles. Figure 62 illustrates constraint conditions compared to a normal condition for each kinematic variable of the lower extremity.

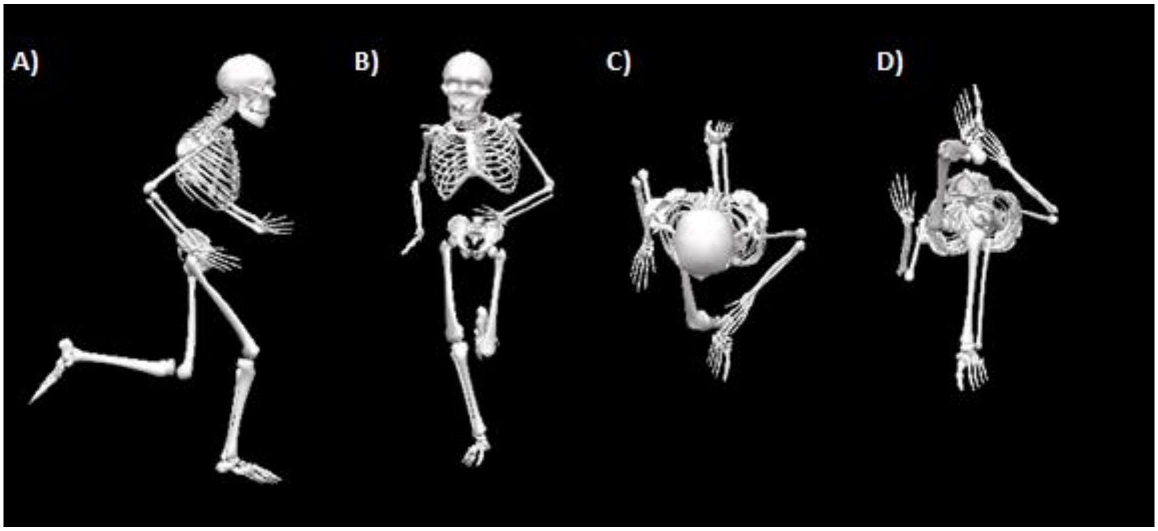

Figure 58. Model View of Athlete at Initial Contact of the Right Foot (A – Right side view, B – Frontal view, C – Superior view, and D – Inferior view).

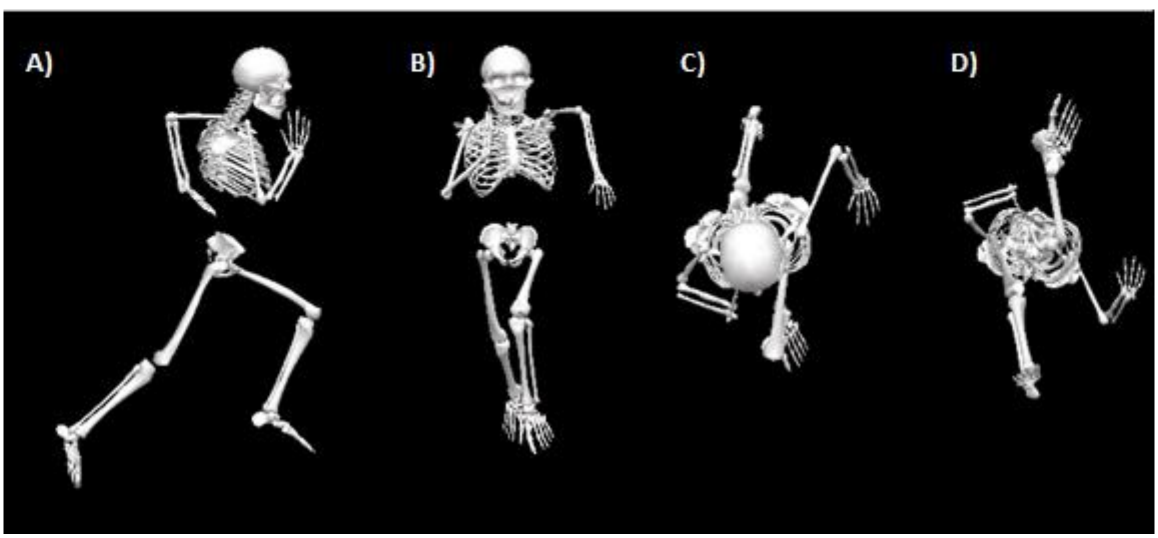

Figure 59. Model View of Athlete at Toe Off of the Right Foot (A – Right side view, B – Front view, C – Superior view, and D – Inferior view).

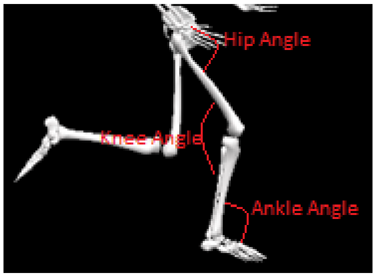

Figure 60. Model View of Ankle, Knee and Hip Angles.

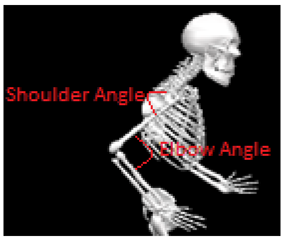

Figure 61. Model View of Shoulder and Elbow Angles.

Figure 62. *Lower Extremity Joint Position.*

Kinematic Variable	CUA	CFA	CH
AA_{IC}	+	+	-
KA_{IC}	-	-	-
HA_{IC}	+	+	-
AA_{TO}	+	+	+
KA_{TO}	+	+	-*
HA_{TO}	-	+	-*

Note. + = Joint kinematic variable greater than N condition (greater knee angle, greater hip flexion and extension). - = Joint kinematic variable smaller than N condition (smaller knee angle, smaller hip flexion and extension). * = Significant difference from N condition.

Lower Extremity

There have been several studies that have identified the changes in ankle, knee and hip angle during arm constrained locomotion (Frere et al., 2009; Miller et al., 2009; Umberger et al., 2008), however to the author's knowledge, no previous research has examined the effects of constraints nor considered the influence of lower extremity constraint on lower extremity joint kinematics. Umberger and colleagues (2008) concluded lower extremity joint angles were nearly identical between arm constraint and non-constraint conditions. Yet it should be noted that this research was completed while individuals were walking at a constant velocity. While investigating athletes running, Frere et al. (2009) noted hip and knee flexion decreased through the sprint with arm constraints. However, Miller and colleagues (2009) demonstrated peak hip flexion increased by 1.7° and peak knee flexion increased by 4° when arms were constricted while running at a constant velocity. This would suggest that the legs are compensating for the lack of arm motion. Furthermore, increases in plantar flexion have been observed with increases in velocity (Kyrolainen et al., 1999). These conflicting findings and the absence of the consideration of leg constraints in the literature has led to the investigation of these variables.

In the present study, there were no significant differences found between constraint conditions and lower extremity joint positions at the ankle, knee and hip at initial contact, (AA_{IC}, KA_{IC} and HA_{IC}) (Figure 20, Figure 22 and Figure 24). While there were no significant differences between the non-constrained or normal condition and any of the constrained conditions, it is interesting to note that the AA_{IC} was the greatest in the CH condition. A larger AA_{IC} is indicative of less dorsiflexion and can be explained by the shorter step length noted during the CH condition. Last, the hip flexion/extension of the hip was smaller during the CH condition, indicating that the constraint did limit range of motion, though not significantly.

Ankle, knee and hip angle of the IL leg were also considered at toe off (TO). No significant difference was noted for the ankle angle between conditions (AA_{TO}) (Figure

21). Interestingly, CUA possessed the greatest plantarflexion at toe off, followed by the CFA condition suggesting that elbow motion contributes to alterations in lower extremity mechanics. These results also suggest that the lack of hip extension did not allow the ankle to be used as an efficient lever and may be the reason for the decrease in non-support time.

Investigation of knee and hip angle did yield significant main effects at toe off, (Figure 24 and Figure 26). KA_{TO} was significantly smaller at toe off during CH when compared to the unconstrained condition, N. KA_{TO} for all conditions were the following: N (163.85°), CUA (165.65°), CFA (164.68°) and CH (161.54°). This suggests that the lack of hip extension did not permit for usual knee extension and was a contributor to decreases in non-support time and VEL. In addition, the HA_{TO} was the smallest during the CH when compared to the other conditions, indicating less hip extension during toe off. Values for hip angle at toe off are the following: N (19.86°), CUA (19.25°), CFA (19.96°) and CH (15.64°). This finding confirms that the hip constraint methodology was successful and further explains the noted decreases in VEL during the CH condition.

Figure 63. *Pelvic Girdle and Trunk Position.*

Kinematic Variable	CUA	CFA	CH
$APPGR_{IC}$	-*	-	-*
$TVPGR_{IC}$	-	-	-
$LPGR_{IC}$	-	-*	-
TF_{IC}	-	+	-
TR_{IC}	+	+	+
LTF_{IC}	-	-	-
$APPGR_{TO}$	-	+	-
$TVPGR_{TO}$	+	-	-
$LPGR_{TO}$	-	+	+
TF_{TO}	-	+	-
TR_{TO}	+	-	+
LTF_{TO}	-	+	+

Note. + = Joint kinematic variable larger than N condition (greater anterior pelvic girdle rotation, transverse pelvic girdle rotation, lateral pelvic girdle rotation, trunk flexion, lateral trunk flexion and trunk torsion). - = Joint kinematic variable smaller than N condition (smaller anterior pelvic girdle rotation, transverse pelvic girdle rotation, lateral pelvic girdle rotation, trunk flexion, lateral trunk flexion and trunk torsion). * = Significant difference from N condition.

Pelvic Girdle Position

The pelvic girdle functions to support the weight of the upper body through the vertebral column, provides an attachment site for musculature in the lumbopelvic-hip complex (LPHC) and transmits ground reaction forces to the vertebral column during gait (Lippert, 2011). The pelvis moves in all 3 planes of motion during bipedal locomotion. Literature investigating pelvic girdle angle in bipedal locomotion is not as common as other joints in the LPHC (Novacheck, 1998; Ounpuu, 1990). Figure 63 illustrates constraint conditions compared to an unconstrained condition for each kinematic variable of the pelvis and trunk.

In the present study, pelvic girdle position in the 3 planes of motion was analyzed during initial contact and toe off of the IL foot ($APPGR_{IC}$, $APPGR_{TO}$, $TVPGR_{IC}$, $TVPGR_{TO}$, $LPGR_{IC}$ and $LPGR_{TO}$). There were significant differences between CUA and CH compared to N on $APPGR_{IC}$ (Figure 26) and $LPGR_{IC}$ (Figure 30) only. $APPGR_{IC}$ data in all conditions were the following: N (30.85°), CUA (28.19°), CFA (30.74°) and CH (26.94°). This finding suggests that the body was re-aligned as a result of the presence of the hip and upper arm constraint. The significant difference noted for the CUA but not the CFA, suggests that the elbow motion without shoulder motion altered the ability of the body to maintain a more natural alignment. In addition, by preventing hip extension, the body could not adopt a more forward leaning posture, which is more beneficial to sprinting, further explaining why the CH condition yielded the smallest velocity. $LPGR_{IC}$ for each condition were the following: N (5.94°), CUA (3.68°), CFA (.13°) and CH (4.11°). CFA demonstrated a significantly smaller IL $LPGR_{IC}$, providing support for the hypothesis that arm motion effects pelvic girdle motion in the frontal plane and suggesting that arm motion contributes to elevating the pelvis.

There were no significant differences found between conditions for $APPGR_{TO}$ (Figure 27), $TVPGR_{IC}$ (Figure 28), $TVPGR_{TO}$ (Figure 29) and $LPGR_{TO}$ (Figure 31). These findings were surprising as it was anticipated that the constraints, particularly the arm constraints, would have the greatest impact on the transverse pelvic girdle rotation. Upon further consideration, it is hypothesized that the trunk and pelvis rotate in near synchronicity in the transverse plane during sprinting. Since the pelvic girdle rotation was measured with respect to the trunk position, synchronized movements would have yielded few differences.

Trunk Position

The trunk serves as the superior portion of the LPHC. The trunk moves in all 3 planes of motion during bipedal locomotion. Limited literature investigating trunk angle within bipedal acceleration is available (Atwater, 1982). Atwater (1982) concluded that the trunk flexes during acceleration and progressively extends, or becomes more upright, as velocity increases. However, to the author's knowledge, there have been no previous studies involving the effects of arm/hip constraints on trunk position during bipedal acceleration.

In the current study, trunk position in the 3 planes of motion was analyzed during initial contact and toe off of the right foot (TF_{IC}, TF_{TO}, LTF_{IC}, LTF_{TO}, TR_{IC} and TR_{TO}). Surprisingly, there were no significant differences found between conditions and all 6 trunk position measures, TF_{IC}, TF_{TO}, LTF_{IC}, LTF_{TO}, TR_{IC} and TR_{TO} (Figure 32, Figure 33, Figure 34, Figure 35, Figure 36, Figure 37, respectively), particularly in light of the significant findings of pelvic girdle position. It is hypothesized that the lack of significance is due to the rather large standard deviations associated with these measures.

Figure 64. *Upper Extremity Joint Position.*

Kinematic Variable	CUA	CFA	CH
SA_{IC}	-	-	+
EA_{IC}	+	+	+
SA_{TO}	-*	-*	-
EA_{TO}	+*	+*	+

Note. + = Joint kinematic variable greater than N condition (greater shoulder flexion and extension, and greater elbow angle). - = Joint kinematic variable lesser than N condition (smaller shoulder flexion and extension, and smaller elbow angle). * = Significant difference from N condition.

Upper Extremity

To the author's knowledge, no previous research has investigated the effects of constraint on upper extremity joint position during acceleration. Figure 64 illustrates constraint conditions compared to a normal condition for each kinematic variable of the upper extremity. In the present study, there were no significant differences found between conditions at the upper extremity joint positions of the elbow and shoulder at initial contact, SA_{IC} and EA_{IC}, respectively (Figure 38 and Figure 40). It is important to note that the shoulder angle during the arm constraint conditions were nearly 10 degrees smaller than the unconstrained condition, indicating that the arm constraint was successful in limiting motion at the shoulder. Furthermore, SA_{IC} was greatest during CH, indicating greater shoulder extension occurred at initial contact while the hips constrained, suggesting the arms were swinging more to aid the acceleration of the athlete (Figure 64). Though more research is needed, an increase in shoulder extension in the presence of diminished hip flexion may cast dispersions on the notion that arm swing is simply for the conservation of angular momentum.

There were significant main effects found for the shoulder angle and elbow angle at toe off (Figure 39 and Figure 41). SA_{TO} significantly decreased during CFA when compared to all other conditions demonstrating that the arm constraints provided limitation to shoulder motion (Figure 64). Furthermore, EA_{TO} was significantly greater in all constraint conditions when compared to N (Figure 64). EA_{TO} was smallest in N indicating more flexion when compared to the constraint conditions, however it should be noted that the elbow angle for the N and CH conditions yielded very similar results, suggesting that the motion of the forearm was not affected by the hip constraint.

Summary

The joint kinematic data from the current study did not yield anticipated results. There were significant differences in the following variables: KA_{TO}, HA_{TO}, $APPGR_{IC}$, $LPGR_{IC}$, SA_{TO}, and EA_{TO}. Results for HA and SA indicate that the arm and leg constraints completed the purpose of limiting motion at the respective joints. A significant decrease in KA_{TO} in the CH condition could have contributed to the diminished AA_{TO} noted. If knee angle was smaller, suggesting greater flexion, greater dorsiflexion may be noticed at the. Moreover, SL_L was significantly lower in the CH condition when compared to the N condition suggesting that the greater flexion noted in CH, decreased the ability to produce longer steps thus decreasing ST_L and VEL. Essentially, the decrease in KA_{TO} during the CH condition may suggest that knee angle affected ST_L and SL_L and thus VEL.

Furthermore, with the significant decrease in $APPGR_{IC}$ during the CUA condition and $LPGR_{IC}$ during the CFA condition, it is suggested that arm swing is related to pelvis rotation, though more research is needed to identify the implications for these results. Additionally, the significant increase in pelvic girdle motion may have decreased the SL_L that was noted in the CUA and CFA conditions. While SL_L decreased and ST_L increased for these conditions, VEL significantly decreased suggesting that the decrease in arm motion caused an increase in pelvic girdle motion which affected step length and contact time. Additional support for this claim will be noted while investigating LD activity in Section 4. Moreover, a significant difference was noted in SA_{TO} in CUA and CFA. This further supports that the arm constraints completed the purpose of limiting motion at the shoulder joint.

Section 3: N, CUA, CFA and CH Conditions on Gait Kinetics

This section will discuss the results regarding the influence of constraint conditions CUA, CFA and CH on ground reaction forces and impulses during right foot contact in comparison to the N condition. This section includes the results on peak vertical ground reaction force ($VGRF_{PEAK}$), time to peak vertical ground reaction force ($VGRF_{TTP}$), vertical ground reaction impulse (VGRI), peak braking ground reaction force ($BGRF_{PEAK}$), braking ground reaction impulse (BGRI), peak propulsive ground reaction force ($PGRF_{PEAK}$), and propulsive ground reaction impulse (PGRI). Figure 65 illustrates constraint conditions compared to a normal condition for each kinetic variable.

Figure 65. *Ground Reaction Forces and Impulses.*

Kinetic Variable	CUA	CFA	CH
$VGRF_{PEAK}$	-	-	-
$BGRF_{PEAK}$	-	-	-*
$PGRF_{PEAK}$	-	-	+
$VGRF_{TTP}$	-	-	+
VGRI	+	+	+
BGRI	-	-	-*
PGRI	-	-	+

Note. + = Kinetic variable larger than N condition (greater peak vertical, braking and propulsive ground reaction force and greater vertical, braking and propulsive ground reaction impulse). - = Kinetic variable smaller than N condition (smaller peak vertical, braking and propulsive ground reaction force and smaller vertical, braking and propulsive ground reaction impulse). * = Significant difference from N condition.

Peak Vertical, Braking and Propulsive Ground Reaction Forces

To increase horizontal velocity, an athlete must have the ability to increase the amount of force applied to the ground. Several studies have investigated ground reaction forces during locomotion (Bohn et al. 1998; Hunter et al., 2010; Weyand et al., 2000; Young, 2007) yet only a few studies have investigated ground reaction forces during acceleration (Merni et al., 1992; Mero, 1988; Hunter et al., 2005). General findings indicate that individuals running at higher velocities exhibit a larger VGRF. Bohn and colleagues (1998) stated that more powerful acceleration is evident in the sprinter who produced greater VGRF and PGRF. In addition, it has been noted that greater PGRF yields faster sprint times (Merni et al., 1992; Mero, 1988). Interestingly, while investigating the effects of arm swing constraints on running at a constant velocity, Miller and colleagues (2009) observed a decrease in $VGRF_{PEAK}$. However, to the author's knowledge there have been no previous studies involving the effects of arm/hip constraints during bipedal acceleration.

In the present study, there was no significant difference found between constraints for $VGRF_{PEAK}$ (Figure 42). However, $VGRF_{PEAK}$ was the greatest in N when compared to the other conditions (Figure 65) supporting previous work which has indicated that larger $VGRF_{PEAK}$ yields larger velocity. Although, differences between constraints were not significant, the results indicate $VGRF_{PEAK}$ decreases with greater arm constraints, similar to previous research.

On the other hand, there were significant differences between conditions and $BGRF_{PEAK}$ (Figure 45). Constraints led to a significant decrease in $BGRF_{PEAK}$ when compared to N (Figure 65). $BGRF_{PEAK}$ for all the conditions were the following: N (488.86 N), CUA (453.57 N), CFA (422.17 N) and CH (312.99 N). The lowest value occurred during the CH condition suggesting braking forces were decreased due to the limited hip motion. This is consistent with the decreased step length, more plantarflexed position and lower velocity noted for the CH condition.

There was no significant main effect found between constraint conditions and N on $PGRF_{PEAK}$ (Figure 47). Interestingly, CH produced the greatest amount of $PGRF_{PEAK}$ which was unexpected to the author (Figure 65). It has been noted that greater $PGRF_{PEAK}$ yields higher velocities (Hunter et al., 2005); however CH yielded the lowest velocity. Additionally, there was no significant main effect between constraint and $VGRF_{TTP}$ (Figure 43). However, it must be noted that CH took the longest time to reach $VGRF_{PEAK}$ after initial contact while CFA acquired $VGRF_{PEAK}$ after initial contact the quickest. Furthermore, KA_{TO} significantly increased with arm constraints and an increase in flexion was observed at HA_{IC}. Additionally, an increase in plantarflexion at AA_{TO} was observed with arm constraints when compared to the N condition. It is proposed that with the lack of effective arm swing, the lower extremity adapts to this change and adjusts the acceleration and position of the joints to further enhance acceleration and velocity.

Vertical, Braking and Propulsive Ground Reaction Impulses

There have been previous studies that have investigated ground reaction impulses during bipedal acceleration (Hunter et al., 2005; Mero, 1988); however to the author's knowledge no previous research has investigated the effects of arm and leg constraints on acceleration, with regards to ground reaction impulses. Mero (1988) stated that propulsive ground reaction impulse (PGRI) significantly correlates with sprint velocity concluding that as PGRI increased, velocity increased as well. In addition, Hunter and colleagues (2005) examined acceleration at the 16 m mark and revealed the faster athlete produced greater PGRI.

In the present study, there was no significant difference found between conditions on VGRI (Figure 44). On the other hand, a significant main effect was found for BGRI (Figure 46). N demonstrated the greatest BGRI when compared to constraints (Figure 65). Values for BGRI were the following: N (14.61 Ns), CUA (13.51 Ns), CFA (12.31 Ns) and CH (7.08 Ns). Specifically, CH yielded the lowest BGRI, which suggests a lack of hip motion decreases velocity, step length, and causes a more plantarflexed foot, all of which decreases the impulse in the anterior direction. Furthermore, there was no significant difference found between conditions on PGRI (Figure 48).

Summary

Several of the results from the kinetic measures during arm and leg constraints were unexpected. Significant differences were found with variables $BGRF_{PEAK}$ and BGRI. Specifically, condition N produced the greatest $BGRF_{PEAK,}$ while demonstrating the greatest VEL and SL_L and smallest ST_L across all 4 conditions. It was suggested that limited arm motion produces more intense lower extremity motion to continue acceleration. However, CUA and CFA produced significantly lower VEL showing that the adaptations that were made were not sufficient enough to recover the limitation of lack of arm swing.

CH produced the lowest BGRI and the greatest PGRI, but the smallest velocity. This result is very difficult to explain as a greater propulsive impulse should yield a greater change in momentum, which was not realized here. Further investigation is needed to determine if the influence of the ground reaction forces are diminished, absorbed or attenuated as a result of the hip constraint. That is to say, does the hip constraint prevent the body from utilizing the ground reaction forces to yield the expected mechanical outcome of the ground reaction force data?

Section 4: N, CUA, CFA and CH Conditions on Muscular Activity

This section will discuss the results regarding the influence of constraint conditions on the muscular activity of the latissimus dorsi (LD) and the gluteus maximus (GM). This section includes the discussion of the results of the N, CUA, CFA and CH conditions on peak activity of the IL LD at initial contact and toe off, peak activity of the contralateral (CL) LD at initial contact and toe off, peak activity of the IL GM at initial contact and toe off and peak activity of the CL GM at initial contact at toe off ($ILLD_{IC}$, $ILLD_{TO}$, $CLLD_{IC}$, $CLLD_{TO}$, $ILGM_{IC}$, $ILGM_{TO}$, $CLGM_{IC}$ and $CLGM_{TO}$, respectively). All sEMG data are represented as peak values of a percentage of the MVIC. Figure 66 illustrates constraint conditions compared to a normal condition for each sEMG variable.

Figure 66. *sEMG data of the LD and GM.*

Electromyography Variable	**CUA**	**CFA**	**CH**
$ILLD_{IC}$	+	+	+
$CLLD_{IC}$	-	-	+
$ILGM_{IC}$	+	+	-
$CLGM_{IC}$	-	-	-
$ILLD_{TO}$	-	-	+*
$CLLD_{TO}$	+	-	+
$ILGM_{TO}$	+	+	-
$CLGM_{TO}$	-	+	-*

Note. + = sEMG value larger than N condition (greater peak muscle activation). - = sEMG value smaller than N condition (smaller peak muscle activation). * = Significant difference from N condition.

Latissimus Dorsi

Of the 29 muscles attaching to the LPHC, the LD, is the largest muscle on the back and serves a very important role to static posture and mobility (Scache et al., 1999). Interestingly, the LD is the only muscle of the LPHC that crosses the shoulder joint. The LD has been shown to assist pelvic girdle and trunk motion and possibly affect the techniques of the upper extremity in the performance of athletes (Kendall, 2005; Patel et al., 2012; Vleeming & Stoeckart, 2007). Few studies have examined LD activity in locomotion (Cappellini et al., 2006; Ivanenko, 2004; Novacheck, 1998), however to the author's knowledge; no previous research has investigated LD activity during acceleration. During constant velocity locomotion on a treadmill, Cappellini and colleagues (2006) noted that LD activation was greater as speeds increased between walking and running. In addition, Ivanenko (2004) also reported increases in LD activation with increases in constant velocity, however similar to the previously mentioned study; data were captured during constant velocity locomotion on a treadmill. Interestingly, Novacheck (1998) suggested that externally rotating the arm at the top of arm swing during sprinting stretches the LD and aids in the creation of potential energy to assist proper posterior pelvic girdle rotation, which is essential for performance at constant maximal velocity. While these studies provide valuable insight into the muscle activity of the LD, to the author's knowledge, no research has attempted to evaluate the influence of acceleration and arm and leg constraints on LD and GM activation.

The discussion of sEMG data on bilateral LD and GM will be ordered such that all sEMG data at initial contact (Figure 57) will be discussed first followed by sEMG data at toe off (Figure 58). In the present study, $ILLD_{IC}$, $CLLD_{IC}$, $ILGM_{IC}$, and $CLGM_{IC}$

were observed at initial contact. Neither $ILLD_{IC}$ nor the $CLLD_{IC}$ demonstrated any significant difference but the CH condition did yield the greatest peak, suggesting a relationship between hip movement and LD activity. It is suggested that the increase in LD activity may be due to the legs having to generate more force in an attempt to overcome the constraint. While the legs were attempting to generate greater force to overcome the constraint, the CL LD increased activity due to the cross effect between CL GM and IL LD. It was apparent that when the CL GM increased activity, the IL LD increased activity as well, supporting the cross effect between these muscles (Myers, 2009; Vleeming & Stoeckart, 2007).

In the present study, $ILLD_{TO}$, $CLLD_{TO}$, $ILGM_{TO}$, and $CLGM_{TO}$ were also observed at toe off. There was a main effect for constraint condition on $ILLD_{TO}$ (Figure 50). Peak sEMG data for $ILLD_{TO}$ were the following: N (1.94 %MVC), CUA (1.82 %MVC), CFA (1.8 %MVC) and CH (3.21 %MVC). CH produced greater $ILLD_{TO}$ then the N condition, suggesting an increase in LD activity was necessary to support the limited hip motion, as was noted in the $CLLD_{IC}$. In addition, there were no significant differences found for the $CLLD_{TO}$ variable between conditions (Figure 52). However, similar to other LD measures, the greatest activation was observed in the CH condition.

Gluteus Maximus

The GM is a component of the LPHC and is found on the posterior aspect of the pelvis, with responsibility for extending and externally rotating the hip as well as contributing to posterior pelvic girdle rotation (Floyd, 2009). Several studies have identified electromyography (sEMG) patterns of the GM increasing during acceleration when compared to constant velocity sprinting (Kyrolainen et al., 1999; Mero & Peltola, 1989). In particular, Mero and Peltola (1989) found that sEMG patterns of the GM increase during acceleration when compared to constant velocity running. In addition, a study completed by Kyrolainen et al. (1999) concluded that during foot contact, there was a 4.8% increase in sEMG activity of the GM during the acceleration phase when compared to the maximal velocity phase. While these studies have advanced the understanding of the muscle mechanics of running, to the author's knowledge this is the first study to investigate the influence of acceleration and arm and leg constraints on GM activation.

There were no significant differences found between conditions for $ILGM_{IC}$ (Figure 53) however; CUA and CFA produced the two greatest activations while CH generated the least activation in $ILGM_{IC}$. The arm constraint conditions experienced an increase in GM activation during initial contact suggesting the legs had to generate more force to enhance acceleration due to the lack of arm swing. Additionally, it is suggested that the increase in activity seen in the GM may have been due to the longer step lengths seen in CUA and CFA when compared to CH.

There were no significant differences found between conditions for $CLGM_{IC}$ (Figure 55). N produced the greatest activation of the $CLGM_{IC}$, followed by CUA, CFA and CH producing the least activation of the CL GM. Analogous to the increase in activity of the IL and CL LD, the IL and CL GM produced less activation in the CH condition. Essentially, while increases were observed in LD activity, decreases were observed in GM activity as well in the CH condition further supporting the cross-effect between these muscles. Furthermore, there was no significant difference found between conditions on $ILGM_{TO}$ (Figure 54). Interestingly, the two arm constraint conditions yielded the greatest $ILGM_{TO}$ further supporting the notion that there is a connection between the arms and pelvis, which was noted in Section 2 which discussing the results of joint kinematics. Pelvic girdle motion increased in the CUA and CFA conditions further noting the connection between arm motion and pelvic girdle motion.

Contrastingly, a significant main effect was noted for the CL GM during toe off ($CLGM_{TO}$) (Figure 55). Peak sEMG data for $CLGM_{TO}$ were the following: N (2.17 %MVC), CUA (2.17 %MVC), CFA (2.32 %MVC) and CH (1.59 %MVC). CH yielded significantly lower peak sEMG than the unconstrained condition (N), in addition to the lowest sEMG value when compared to the arm constraint conditions as well (CUA and CFA). Moreover, the greatest activation of $CLGM_{TO}$ occurred in the CFA condition, suggesting that the CLGM was developing greater activation to counteract the decrease in LD activity and arm swing. Though not considered statistically, the $ILLD_{IC}$ and $CLGM_{IC}$ were active together, as were the $CLLD_{IC}$ and $ILGM_{IC}$. This pattern was also noted during the toe off event as well further supporting the cross effect between CL LD and GM and transfer of activation within the thoracolumbar fascia (TLF).

Summary

The main finding of the electromyography data was the cross relationship between CL musculature. For example, while the IL LD was active, the CL GM was active as well. The same occurred on the opposite side. The author concludes that the cross relationship can be supported through the current data; however additional research is needed to validate the relationship with the TLF and these muscles. In addition, LD activity was greater in CH with all the LD variables that were measured ($ILLD_{IC}$, $ILLD_{TO}$, $CLLD_{IC}$, $CLLD_{TO}$), however it must be noted that not all increases were significant. The results of LD activity must also be considered in light of the variables presented in in Section 3 of this chapter. A decrease in arm motion increased pelvic girdle motion suggesting that the LD may have an effect on pelvic girdle motion. A decrease in arm motion with increased pelvic girdle and an increase in LD activity support the findings of Kendall (2005), Patel et al. (2012) and Vleeming & Stockart (2007) that the LD has a role at the pelvis as well. In essence, the LD does more than shoulder motion, an increase in LD activity in the presence of arm constraint suggests that the LD plays a larger role than just arm motion. While LD activity was observed in the arm constraint conditions, it is suggested that since the LD did not provide shoulder motion, the LD altered the pelvic girdle angle due to the distal attachment on the pelvic girdle. Additionally, with the increase in LD activity seen in the CH condition, it is mentioned that the SA_{IC} was greater when compared to the N condition. This suggests that the increase in LD activity provided the shoulder extension at initial contact, as opposed to the pelvic girdle motion noted in the arm constraint conditions.

Furthermore, IL and CL GM decreased during the CH condition. Suggesting that the constraints completed their role in limiting hip motion and the increase in LD activation occurred to assist the lack of activation of the GM. The lower activation of the GM coincided with the decrease in SL_L, $VGRF_{PEAK}$, HA_{IC} and HA_{TO} and increase in ST_L, which are all causes for the decrease in VEL observed in CH compared to the unconstrained N condition. With the lack of GM activity to complete motion at the hip, the variables previously stated were affected negatively. The importance of the GM during acceleration can be supported through results from the current study. A decrease in GM activity negatively affects certain spatial-temporal measures which are beneficial for acceleration. It is surprising that with the decrease in GM activity, a greater $PGRF_{PEAK}$ was noted with the CH condition, more research is needed to complete this connection. Additionally, more research is necessary to providing the link between GM activity and additional musculature involved with acceleration.

Section 5: Summary & Future Research

The final section will discuss the conclusions drawn from the current study and directions for future research. The goal of this study was to determine the role of the LPHC during the initial acceleration phase of a sprint and to investigate the effects of constraint conditions on the spatial-temporal kinematics, joint kinematics, gait kinetics and musculature activity during bipedal locomotion acceleration.

The athlete was constrained at either the upper arm (CUA), full arm (CFA) or thigh (CH) while accelerating, and there were several variables that changed significantly from a normal, unconstraint condition (N). Specifically, SL_{L}, CT_{L}, VEL, KA_{TO}, HA_{TO}, $APPGR_{IC}$, $LPGR_{IC}$, SA_{TO}, EA_{TO}, $BGRF_{PEAK}$, BGRI, $ILLD_{TO}$, and $CLGM_{TO}$ all yielded statistically significant differences between constraint conditions and the N condition. Spatial-temporal measures of SL_{L}, CT_{L} AND VEL all decreased during the constraint conditions when compared to the N condition. This study endeavored to contribute to the current understand of running by determining the role of the shoulder, elbow and hips during acceleration. In addition, the role of the LD and GM during acceleration was also investigated. As was expected, constraints diminished VEL; however, elbow motion

(CUA) provided enough impetus to achieve a non-significant difference in VEL in comparison to the unconstrained condition. Furthermore, the increase in arm motion in the presence of hip constraint suggests that the arms do more than conserve angular momentum. Though more research is needed, this project helps to reveal other purposes of arm motion during running. Specifically, since the LD was significantly more active during both shoulder constrained conditions suggest that the LD is doing more than moving the shoulder during running or may imply that the LD must work harder because it is in a less than optimal length to fulfill the purpose of the muscle during acceleration.

The present study also advances the literature by shedding further light on the cross relationship between the LD and GM muscles. That this relationship exists is not new, however the changes brought about by the constraints advances understanding Future research is needed to more fully understand this cross relationship in light of the TLF, which connects the upper and lower extremities through the superficial portion of the LPHC (Myers, 2009). Specifically, the LD and GM are coupled via the TLF and can provide tension to the surrounding structures and the TLF has been noted to aid the LD and GM in providing a pathway for energy transmission in the LPHC (Vleeming & Stoeckart, 2007). However, further investigation is required to completely understand the combined role of the LD, GM and TLF and how each contributes to tension development during human motion.

Future Research

The author suggests that the present study is a novel topic investigating the effect of arm and leg constraints on bipedal acceleration. Future investigations should consider additional musculature of the LPHC during acceleration to better understand the synergist relationship between contralateral muscles. In addition, the effects of arm and leg constraint on walking and constant velocity running need to be investigated to fully understand the changes that are acquired in the presence of constraints, for example, kinematic changes that occur during talking on a cell phone should be considered. Training studies involving the upper extremity, specifically the LD, would be beneficial to understanding the overall effects of the LD during sprinting since much of the previous research completed on acceleration have focused primarily on lower extremity strengthening. In particular a training study could be implemented to determine if a solely training the LD increases velocity and/or acceleration.

As stated earlier, further research is needed to more fully understand the relationship between contralateral muscles during locomotion. Results from the current study imply that there is a cross relationship between CL muscles, however research beyond the LD and GM is required to understand the TLF and additional fascial contributions between musculature. It has been noted that during running when the athlete fatigues, trunk and upper extremity motion may slightly increase to support the lower extremity motion (Elliot & Ackland, 1981). As a result, the effects between contralateral muscles should be investigated to understand how these motions counter the mechanical effects of fatigue.

Last, more realistic constraints should be investigated. For example, locomotion with a handheld music device, dog leash and stroller should be investigated. Furthermore, other types of constraint such as body and arm load carriage, particularly as encountered by the military should be considered. In addition, the foot architecture should be considered to more fully understand the influence of constraints.

Individuals who are focused on accelerating to produce the greatest velocity must understand what occurs within the body to adapt to different conditions during acceleration. The results of the present study suggest that elbow motion is important and arm motion may do more than simply help to satisfy angular momentum. In addition, this project suggests that not only do the LD and GM play a significant role in acceleration, but also work together. To run more efficiently this relationship should be exploited and training protocols should be developed to reinforce this neurological pattern. While more study is needed, this project has made great strides in understanding the mechanics of human acceleration.

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Appendix A

Participant Screening Questionnaire

Please read each question carefully and answer honestly. If you do not understand the question, please ask the investigator for clarification. Check the appropriate answer.

Participant Number: ___________

Yes **No**

___ ___ 1. Are you younger than the age of 19 or older than the age of 30?

___ ___ 2. Do you currently have a musculoskeletal illness that prevents you from performing a 10 meter sprint?

___ ___ 3. Have you had any musculoskeletal injury or surgery within the last year?

___ ___ 4. Do you have any reason to believe that your participation in this investigation may put your health or well-being at risk?

___ ___ 5. Are you capable of performing a 10 meter sprint with no pain or range of motion issues?

___ ___ 6. Are you allergic to adhesives?

___ ___ 7. Are you currently taking any medication affecting balance, alertness or the musculoskeletal system?

Signature of participant ____________________________ Date__________________

If any questions have been selected as “Yes”, the participant will be excluded from the study.

Appendix B

Institutionally Approved Informed Consent Document

INFORMED CONSENT FOR
"The Role of the Lumbo-Pelvic-Hip Complex in Bipedal Acceleration"

You are invited to participate in a research study to investigate the role of the lumbo-pelvic-hip complex in bipedal acceleration. The study is being conducted by Jay Patel, Doctoral Student, under the direction of Dr. Wendi Weimar, Associate Professor, in the Auburn University Department of Kinesiology. With your help it is hoped that the role of the lumbo-pelvic-hip complex is affected during arm and leg constraints during bipedal acceleration. You were selected as a potential participant because you are a male between 19 and 30 years of age, and your health condition and mobility might, through pre-screening health questionnaire to follow, permit you to perform the test safely and successfully. The subject population is that of convenience and not representative of any working population in general.

Purpose: The purposes of this research are: 1) to determine the exact role of the lumbo-pelvic-hip complex during bipedal acceleration under different arm and leg constraints; and 2) to determine the overall effects of different arm and leg swing constraints during bipedal acceleration associated with different arm and leg constraints. The results will aid coaches and athletes to better prepare for all of their athletic endeavors.

Methodology: The testing session will last approximately 150 minutes, over two days. Day one will be approximately 30 minutes, while day 2 will be 120 minutes. Day 1 will be a familiarization day where you will complete the sprints in each condition with no data collection. Data collection will begin on day 2. You will be asked the day of the trials about any current illness or medication taken that could affect your alertness, balance or overall ability to safely perform the sprints. Once electromyography electrodes have been attached to your latissimus dorsi and gluteus maximus muscles, 34 retro-reflective markers will be attached to you, via plastic tape, to complete the 3D Plug-In-Gait model for motion capture. You will perform 3 trials of a 10 meter sprint during 4 randomized arm and leg constraint conditions while being required to use the same shoes through all trials. In addition, you will not wear any upper body apparel. The first condition will be a normal sprint. The second condition will be with your upper arms constrained via ACE wrap and plastic wrap. The third condition will have your upper arm and forearm constricted the same way. Furthermore, the fourth condition will have your legs constrained to limit motion at the hip.

Participant's Initials___________

Your pelvis will be free to move in all 4 conditions. Each trial will be separated by 4 minutes of rest and conditions will be separated by 10 minutes of rest and preparation. Kinematic, kinetic and electromyographic data will be collected during all the trials.
Risk: It is possible that you may sustain muscle soreness or a muscle injury. In the unlikely event that you sustain an injury from participation in this study, you will be required to assume full financial responsibility for your own medical care. Participants are responsible for any and all medical cost resulting from injury during the study. Further, you may discontinue participation at any time without penalty.

Benefits: There is no direct benefit to you.

Confidentiality: Any information obtained in connection with this study that can be identified with you will remain confidential. Your decision whether or not to participate will not jeopardize your future relations with Auburn University and the Department of Kinesiology. If you decide later to withdraw from the study you may also withdraw any identifiable information, which has been collected about you, in this study. A copy of this form is for you to keep for your records.

If you have questions about this study, please ask them now or contact Jay Patel at pateljh@tigermail.auburn.edu or Dr. Wendi Weimar at weimawh@auburn.edu. Both can be contacted at 334-844-1468.

If you have questions about your rights as a research participant, you may contact the Auburn University of Human Subject Research of the Institutional Review Board by phone (334) 844-5966 or email at hsubjec@auburn.edu or IRBChair@auburn.edu.

HAVING READ THE INFORMATION PROVIDED, YOU MUST DECIDE WHETHER OR NOT YOU WISH TO PARTICIPATE IN THIS RESEARCH STUDY. YOUR SIGNATURE INDICATED YOUR WILLINGNESS TO PARTICIPATE.

______________________________ _________
Name Date

____
Participant’s signature

______________________________ _________
Signature of Investigator: Jay Patel Date