Trunk Extensor Fatigue Decreases Jump Height Similarly Under Stable and Unstable Conditions with Experienced Jumpers

by

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Contents

List of abbreviations

ANOVA: analysis of variance

COG: center of gravity

CMJ: counter movement jump

EMG: electromyography

ES: effect sizes

GRF: ground reaction forces

Hz: hertz

H Reflex – Hoffman Reflex

iEMG: integrated electromyography

LJ: lateral jump

MAV: mean average voltage

SSC: stretch-shortening cycle

SE: standard error

Kg: kilogram

m: meter

ms: milliseconds

mV: millivolts

s: second

Introduction

Achieving maximal jump height and rapid changes in direction are an integral component of many sports (e.g., basketball, volleyball, gymnastics). The ability to jump higher and cut quicker is of upmost importance when attempting to outperform the competition. Hoffman and colleagues (1996) displayed that Division 1 collegiate basketball players with the highest vertical jump receive more playing time than the players that could not jump as high. In order to maximize athletic movement, we must understand where movement originates. Before movement can occur, the core musculature needs to stabilize the body's center of gravity over the base of support and generate force through the transfer of segmental angular momentum (Kibler et al., 2006). Throughout the literature, there are a variety of descriptions or definitions for the core musculature.

Core musculature

Anatomy of core

The literature provides a variety of definitions and descriptions for the core musculature. Behm and colleagues (2010) conducted a review process to precisely distinguish the different aspects of the core musculature and function. They concluded that the anatomical core consisted of the axial skeleton (which includes the pelvic girdle and shoulder girdles) and all soft tissues (i.e., articular and fibro-cartilage, ligaments, tendons, muscles, fascia) with a proximal attachment originating on the axial skeleton, regardless of whether the soft tissue terminates on the axial or appendicular skeleton (upper and lower extremities). All of these soft tissues and axial skeleton work to move in a variety of eccentric, concentric, or isometric motions (Behm et al., 2010).

Within the core musculature, there are three active muscle subsystems. These three subsystems are divided amongst global, local, and transfer muscle groups. The global core muscles consist of: longissimus thorasic, illicostal thorasic, quadratus lumborum, rectus abdominis, external obliques, and internal oblique. The local muscle system consists of: intertransversaii, interspinalis, multifidus, longissimus lumborum, illiocostal lumborum, quadradus lumborum, transversus abdominus, and internal obliques. The transfer muscles consist of: hip flexors, extensors, abductors, adductors, scapular stabilizers, and muscles that act on the glenohumeral joint (Colston, 2012). The groups work together to provide multi-segmental stiffness over a large range and act as prime movers during dynamic activities (Behm et al., 2010).

The local core musculature (e.g., multifidus, rotators, interspinalis, intertransversalis) is responsible to provide inter-segmental stiffness between adjacent vertebrae. Additional local axial skeleton stabilizers include the transverse abdominis, internal oblique abdominis, quadratus lumborum, diaphragm, and the levator ani. These six muscles work together to increase intra-abdominal pressure. The importance of increasing the intra-abdominal pressure protects the spine from compressive forces and increases core stiffness to reduce perturbations caused from opposing forces acting on the body (e.g., opposing player, impact from landing, a strike from combat sport). Another key function of the core musculature is to coordinate and control movement (Colston, 2012). The global core musculature is the largest mass of the trunk and its main responsibility is to maximize force production. Finally, the axial-appendicular (transfer muscles) work to transfer forces /momentum to the limbs to move or initiate throwing movements.

Core stability

In addition to the anatomy of the core, the concept of core stability also needs to be understood. Once again, there is no universally-accepted definition of core stability. Kibler et al. (2006) defined core stability as "The ability to control the position and motion of the trunk over the pelvis and legs to allow optimum production, transfer, control of force and motion to the terminal segment in integrated kinetic chain activities." The importance of the core musculature in human movement is to provide a foundation for the transfer of angular momentum to the limbs and to maintain balance through postural adjustments to keep one's center of gravity over the base of support (Stranget al., 2009). Pre-programmed muscle activation (feedforward programs) allow for efficient local and distal function.

Dysfunction of core

Furthermore, understanding dysfunction of the core can also provide further insights in terms of performance and health. One function of the core is to stabilize the spine and protect against lower back injuries. Lower back injuries represent 10 to 15% of all athletic injuries in the United States (Colston, 2012). Correspondingly, the pain and discomfort caused by lower back pain affects postural stability and impairs balance (Woon Ham et al., 2010). In addition, there is evidence for proprioceptive deficits among patients with recurrent lower back pain related to postural balance and neuromuscular performance (Woon Ham et al., 2010). In fact, it has been shown that having a strong and endurant core can reduce the likelihood of hip, pelvis, thigh, knee and ankle injuries (Colston, 2012). In essence, it is widely accepted that having a strong and healthy core is of upmost importance for all individuals (Kibler et al., 2006; Behm et al, 2010; Sandrey and Mitzel, 2013).

However, a review conducted by Hibbs and colleagues (2008) suggest that there is a lack of research supporting the idea of core stability and the effects on athletic performance. For instance, when examining the effects of core stability training in experienced runners, Sato and Mokha (2009) could not establish a significant link between core training over six weeks and any improvement in lower limb stability or ground reaction forces. They suggested that the mechanics of running were not affected with experienced runners when core musculature conditioning is improved. Conversely, a study that investigated the throwing velocity of female handball players following a 6-week core stabilizing regime, reported a significant increase in throwing velocity (Saterbakken et al., 2011). This investigation suggested that a high level of core stability and strength is required for generating force in multi-segmental movements.

Stages of core training

When designing or applying a specific training program to an individual, it is important to address individual needs. Depending on the condition and experience of the individual, tapering should be taken into account before jumping into potentially dangerous movements. Colston, (2012.) separated core-training programs into three subgroups

I. Cognitive phase – cognitive-oriented problems: contract deep muscles to increase precision and skill. Essentially, learning how to turn on and off the global core muscles. i.e., practicing abdominal bracing through planks and other stationary exercises.

II. Associative phase – progression to more challenging positions. Improve on consistency of performance, success, and refinement. This would involve the integration of both global and local muscles groups. i.e., medicine ball throws and other dynamic exercises that focus on contraction and relaxation of the core musculature under control.

III. Autonomic phase – tasks become habitual. Concentration is placed on the task at hand and not the activation of the core musculature. i.e., sports specific movements, such as throwing a baseball, kicking a soccer ball, swinging a baseball bat. This phase should incorporate the athlete's ability to increase power and speed of movement by increasing the contribution from the core. It is one thing to work on core endurance and strength, but it takes practice to transfer this additional strength to a sports specific movement.

Hence, the more experienced an athlete is, the less likely they are to be affected by exogenous variables. However, studies that have shown a decrease or no change in core muscle activation in unstable environments typically used a trained group of subjects (Wahl & Behm, 2008; Bressels et al., 2009). The more training experience a group of subjects has, the less likely there will be differences in muscle activation and performance in an unstable environment.

Training the Core

A review conducted by Behm and colleagues (2010) outlined specification for training the core musculature for athletic and non-athletic populations. In terms of training the core, it is important to distinguish between three groups: (i) athletes that want to improve athletic performance, (ii) those that want to maintain health, and (iii) those in need of rehabilitation.

(i) Athletes that are training for maximal strength, power, and velocity of movement should emphasize high-intensity closed kinetic movements such as Olympic lefts, squats, and deadlifts (Behm et al., 2010).

(ii) For fitness and health conscious individuals, ground-based free weightlifts (e.g., back squats, dead lifts, Olympic lifts, and lifts that involve trunk rotation) should form the foundation of exercises to train the core musculature. These closed kinetic chain movements allow for moderate levels of instability that allow for the simultaneous development of upper and lower extremity strength, thereby addressing all links in the kinetic chain (Behm et al., 2010).

(iii) For individuals that are in need for rehabilitation, instability devices are becoming increasing popular in order to reduce load on joints and increase muscle activation. Additionally, such training may enhance co-activation between agonist and antagonist muscles. This allows for stiffening of joint complexes and may enhance rate of recovery of an injury to the core or elsewhere (Behm et al., 2010).

Hence according to the aforementioned review specifications, the use and advantages of core training may be population specific. There are a number of studies demonstrating instability-induced changes in muscle activation with untrained or recreationally trained individuals (Vera-Garcia et al., 2007; Arjmand et al., 2005). On the contrary, there are a number of studies indicating a lack of instability-induced changes in core or trunk muscle activation with highly resistance-trained individuals (Wahl ant Behm, 2008; Bressels et al., 2009). Most of these studies involve slow or moderate speed movements (e.g. bench press, chest press, shoulder press, squats, curl-ups) upon a stationary unstable or stable platform or device. Highly trained athletes rarely move at slow or moderate speeds or remain stationary when competing. Thus, there is a need to expand upon the instability resistance training literature with more dynamic activities especially using experienced and/or athletic populations. In addition, during training or competition, it is quite likely that an athlete will experience fatigue that could affect performance under unstable conditions that might not be evident when tested under rested conditions. As previously mentioned, dysfunction of the core can lead to injuries of the lower back, hip, pelvis, thigh, knee and ankle injuries. Since fatigue reduces force and one's ability to coordinate, it is reasonable to suggest that trunk fatigue will also increase the likelihood of such injuries. Additionally, the unpredictable and unstable surfaces in sport can also increase the likelihood of injuries (e.g. stepping on a foot and rolling one's ankle, running on an uneven field, or contact with another player). Thus, understanding the interaction of fatigue and sport specific ballistic movements on unstable devices warrants further investigation.

Effects of Fatigue

The immediate effects of fatigue include reductions in force and balance. When balance is perturbed, there is a direct correlation with a decrease in athletic performance as well as an increase for the probability of injury (Douris et al., 2011).

Stabilizing and correcting the trunk's posture allows for one to position their center of gravity (COG) over their base of support. When the COG is not over the base of support, there is a loss of balance and movement may not be efficiently transferred in the desired plane of motion. Research in the area of jumping mechanics demonstrated that in order to maximize jump height, that all forces must be transferred into the vertical plane (Bobbert et al., 2011). When postural muscles become fatigued, there is a tendency for an increased postural sway and lack of postural control. These mechanisms may cause a subject to displace their center of gravity in a sub-optimal position and displace ground reaction forces in a more horizontal direction, thus not reaching maximal jump height. Kean and Behm (2006) discovered that CMJ height increased following a 6-week fixed foot balance training regime and speculated that a balance training-induced decrease in postural sway may have resulted in reaction forces being applied in a more vertical direction. This illustrates the importance of balance in jumping and shows that a decrement in balance could cause a decrease in CMJ height. A study by Surenkok and colleagues (2008) used an isokinetic machine to induce trunk fatigue and tested for lactate accumulation as well as dynamic balance test and found a significant positive correlation with lactate buildup and a decrease in dynamic balance. This also agrees with our hypothesis that fatiguing the posterior chain does affect a person's ability to correct posture and maintain balance. In multi-joint movements, the activation patterns typically follow a proximal to distal sequence, especially in locomotion and jumping movements (Kopper et al., 2012). A disparity amongst control and musculoskeletal properties leads to an unbalanced increase in segment angular velocities, causing the concentric velocity of some muscles to be disproportionally high and the total work produced to be unnecessarily small (Bobbert et al., 2011), basically, making a skilled jumper appear to be un-coordinated and unfamiliar with the movement. By inducing trunk fatigue, both balance and momentum transfer will be perturbed and thus, jumping performance should be affected. Although, some investigators have suggested core stability does not play a significant role in athletic performance. For instance, a study that investigated isometric core stability was unable to show a significant correlation to functional dynamic movements (Okada et al., 2011). Not only would trunk fatigue alter segmental mechanics and balance, but the concept of cross-over fatigue can also affect the neuromuscular properties of limb movements.

Cross-over fatigue

Cross-over fatigue is when a working muscle group causes fatigue in non-working tissue. While there are conflicting studies in the literature, a number of studies have documented fatigue of non-target or non-localized muscles. Rattey and colleagues (2005) examined the effects of cross-over fatigue by isometrically fatiguing the dominant leg and proceeded to measure the EMG activity. Following the fatiguing protocol, they observed a decrease in voluntary action as well as a decrease in iEMG activity. They cited that centrally mediated mechanisms may have been the reason for changes in the non-exercised leg. Another study that looked at cross-over fatigue discovered a post-fatigue decline in CNS excitability (Post et al., 2008). Furthermore, another study that investigated the effects of fatiguing the hand flexors using an isometric handgrip contraction showed a temporary decrease in EMG activity in the non-exercised plantar flexor muscles (Kennedy et al., 2012). This article also indicated that the detriment in performance in the plantar flexors was affected by systemic central fatigue. Ipso facto, if fatigue disrupts one's ability to maintain posture, then the effects of the unstable surface (if any) should be magnified by fatigue.

Fatigue of the trunk has been shown in several studies to cause a decrement in balance (Surenkok et al., 2008; Parreira et al., 2013; Vuillerme et al., 2007). All three of these studies investigated trunk fatigue and the effect on postural sway, static and dynamic balance in a stationary position. In many popular athletic competitions (e.g., basketball, soccer, volleyball), little emphasis is put on stationary balance. To the authors' knowledge; there is no study available that examined trunk fatigue and jumping/cutting maneuvers on unstable devices. Furthermore the possible detrimental fatigue effects could be exacerbated when attempted to perform athletic maneuvers on unstable surfaces.

Effects of instability

Lower limb movements that are ballistic in nature (i.e., cutting and jumping) rely on the stretch-shortening cycle for optimal performance. The stretch-shortening cycle (SSC) describes a muscle function in which the pre-activated muscle-tendon complex is lengthened in the eccentric phase preceding the immediate concentric phase (Taube et al., 2012). In humans, the SSC is important for locomotion, hopping, jumping, and throwing motions (Komi, 2000). Basically, the SSC mechanism is important in sport and everyday living. Due to the rapid activation of the SSC, the rigidity of the surface plays a significant role in the performance of SSC movements. Notably, adjustments in leg stiffness affect the efficiency of the SSC when there is a change in the surface stability. Ferris et al., (1998) demonstrated that runners tend to adjust leg stiffness when running on compliant surfaces in order to maintain the center of mass displacement on each stride. This autonomous process increases contact time and feasibly would decrease performance in jumping tasks on unstable surfaces. In cutting, jumping, bouncing, and bounding movements, humans can adjust the actions of the body's many musculoskeletal elements, including muscles, tendons, and ligaments (Farley et al., 1998). The interaction between an individual's leg and surface is similar to a simple spring-mass system. Meaning that, when surface compliance changes, over-all leg stiffness will increase to match the decrease in surface rigidity (Farley, et al., 1998). The autonomous function aims to minimize the change in ground contact time and displacement of the COM.

Thus, understanding how the lower limbs react under duress and fatigue may shed some light on the importance of instability training to enhance performance during jumping and cutting maneuvers. In addition to alterations in leg stiffness and the SSC, deviations in posture have been shown to increase trunk activation. Thus, when a surface becomes more unstable, the greater the postural adjustments that the trunk needs to make in order to keep the body's COG over the base of support. Several studies have shown an increase in trunk stabilizer muscles when comparing stable to unstable environments (Behm et al., 2005; Anderson and Behm, 2005; Vera-Garcia et al., 2000). Alternatively, several studies have shown a significant decrease in trunk stabilizers when performing dynamic movements on an unstable device (Marshall et al., 2006; Freeman et al., 2006; Bressels et al., 2009).

Use of instability devices

The use of instability devices to train the core is becoming increasingly popular despite the conflicting findings in the literature. There are several reasons why coaches, trainers, and practitioners are encompassing instability devices in their rehabilitation and periodization programs. Firstly, researchers have indicated that there is in increase in muscle activation when an individual exercises in an unstable environment (Behm et al., 2005; Bressel et al., 2009). Additionally, Anderson and Behm (2004) found maintenance of EMG activity with a significant decrease in force output when performing closed kinetic chain movements on unstable devices. Although this suggests that the use of instability devices produce no discernible benefits for strength training, there is some logic to using such devices in a rehabilitation setting or de-loading phase. In a rehabilitation setting or de-loading phase of an athletic periodization program, instability training could be beneficial to maintain a high level of muscle activation, while removing compressive forces from the spine and lower limb joints. On the other hand, several studies have shown a decrease in muscle activity when exerting force on unstable surfaces, particularly for lower limb exercises (McBride et al., 2006; McBride et al., 2010; Bressel et al., 2009). This leaves the question whether or not we should use instability devices or should we perform all movements on a stable surface. Anecdotally, it is considerably more difficult to run on ice compared to dry land. Yet it would seem illogical to recommend for competitive runners to train on ice to enhance dry land running simply because it is more difficult. A similar phenomenon exists when using instability devices to train athletes.

The neuromuscular responses have been shown to differ amongst the training status of the investigated group. Notably, some conflicting findings in the literature have been due to the training status of the subjects investigated. Anderson and Behm (2004) demonstrated an increase in activation of the lower limb and trunk musculature when performing squats under unstable conditions. Likewise, Behm and colleagues (2004) found an increase in trunk activation when performing bench and shoulder press on a Swiss ball. All the aforementioned studies as well as other similar studies (Stanforth et al., 1998; Vera-Garcia et al., 2000) have used sedentary, elderly, or recreationally active individuals. As previously mentioned, other studies that have evaluated instability training with individuals who have trained extensively with relatively unstable free weights have shown a decrease in EMG activity (McBride et al., 2010; Bressel et al., 2009).

The biomechanical challenge of moving on an unstable device can replicate the unpredictable nature of sports. Instability in sport can vary from a multitude of sources; surface instability can range from frozen surfaces in winter sports to uneven fields in soccer and in addition to surface instability, athletes also have to cope with perturbations in their center of gravity caused by contact from other players. When an athlete loses balance due to contact from other players or playing surface, the individual can experience decrement in performance and increase the likelihood of injury. The logic behind the decrease in muscle activation is that when you perturb the base of support, you must reduce load to a manageable weight. Thus, the reduction in load reduces EMG activity further than a stable environment with near maximal load. For instance, McBride et al. (2010) showed a decrease in lower limb muscle activity on unstable as compared to stable surfaces. However, these studies reported the effects of surface instability during the performance of isometric and dynamic squats. During cutting (e.g., lateral jumps) and jumping, it seems that EMG activity in lower limb muscles is preprogrammed during the preactivation phase (Dyhre-Poulsen et al., 1991; Avela et al., 1996) and affected by stretching loads (Avela et al. 1996, Komi and Gollhofer 1997, Fleischmann et al., 2010; Hoffrén et al., 2011) during the braking phase. In fact, it has been shown that muscle preactivation is related to the appearance and magnitude of spinal stretch reflexes during ground contact of drop jumps (Avela et al. 1996). In this regard, jumping and landing on unstable/foam surfaces may dampen the impact at ground contact which could reduce both, muscle preactivation and reflex activity.

Conclusion

Throughout the literature, it is evident that there are conflicting findings whether or not the use of instability devices are warranted in training. Many studies have used closed kinetic chain movements and very few have used ballistic movements such as jumping and cutting maneuvers on unstable devices. In explosive sports (e.g., soccer, volleyball, basketball) the surface instability is low to moderate. For this reasons we will use a moderately unstable device for our investigation. The effects of instability are more pronounced with untrained individuals due to cognitive-inhibition (lack of confidence) and may not be due to physiological reasons (Hoffamn et al., 1996). For this reason, we will use experienced jumpers. Additionally in the literature, trunk stabilizer activation has both increased and decreased in activation when moving on an unstable surface. Fatiguing the core should down-regulate muscle activation in core stabilizers. By fatiguing the core, we hope to see a more pronounced difference in performance and muscle activation when comparing surfaces.

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Trunk Extensor Fatigue Decreases Jump Height Similarly

Under Stable and Unstable Conditions with Experienced Jumpers

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Abstract

Purpose: The purpose of this study was to investigate the effects of back extensor fatigue on performance measures and electromyographic (EMG) activity of leg and trunk muscles during jumping on stable and unstable surfaces.

Methods: Before and after a fatigue protocol for the back extensors, countermovement (CMJ) and lateral jumps (LJ) were performed on a force plate under stable and unstable (balance pad on the force plate) conditions. Performance measures for LJ (contact time) and CMJ height and leg and trunk muscles EMG activity were tested in 14 subjects (age: 22.6 ± 5.4 years) during 3 different time intervals for CMJ (preactivation phase, braking phase, push-off phase) and 5 different time intervals for LJ (-30-0 ms, 0–30ms, 30-60ms, 60-90ms, and 90-120ms) in non-fatigued and fatigued condition.

Results: A significant main effect of test (p = .007) was found on CMJ height. CMJ analyses did not show any statistically significant results for jumping performance × surface interactions**.** Additionally, a significant decrease was observed in EMG activation in biceps femoris (BF) and gastrocnemius (GAS) following the fatiguing protocol (p = .008 & p = .03; respectively). LJ contact time was not affected by fatigue or surface interaction. EMG activity was significantly lower in the tibialis anterior (TA) following the fatigue protocol (p = .05).

Conclusion: The present findings suggest that skilled jumpers are not affected by the condition (a moderately unstable surface). Additionally, we observed that the main effect of fatigue negatively impacts CMJ height.

Key Words: countermovement jump, instability, balance, core, lateral jumps,

Introduction

Achieving maximal jump height and rapid changes in direction are an integral component of many sports (e.g., basketball, volleyball, gymnastics). The ability to jump higher and cut quicker is of upmost importance when attempting to outperform the competition. Hoffman~~,~~ et al., (1996) displayed that Division 1 collegiate basketball players with the highest vertical jump receive more playing time than the players that could not jump as high. In addition to maximal jumping ability, athletes are also challenged to perform in both stable and unstable conditions. Instability can vary from a multitude of sources; surface instability can range from frozen surfaces in winter sports to uneven fields in soccer and in addition to surface instability, athletes also have to cope with perturbations in their center of gravity caused by contact from other players. When an athlete loses balance due to contact from other players or playing surface, the individual can experience decrement in performance and increase the likelihood of injury. Thus, understanding the internal mechanisms and exogenous variables that alter jumping mechanics is of great importance to increase performance and reduce the likelihood of injury.

Lower limb movements that are ballistic in nature (i.e., cutting and jumping) rely on the stretch-shortening cycle for optimal performance. The stretch-shortening cycle (SSC) describes a muscle function in which the pre-activated muscle-tendon complex is lengthened in the eccentric phase preceding the immediate concentric phase (Taube et al., 2012). In humans, the SSC is important for locomotion, hopping, jumping, and throwing motions (Komi, 2000). Due to the rapid activation of the SSC, the rigidity of the surface plays a significant role in the performance of SSC movements. Consequently, understanding the interaction between surface instability and the SCC is pivotal when designing training regimes for athletes and the general population to improve performance and to reduce injury.

Notably, a change in surface affects the efficiency of the SSC by adjusting leg stiffness. Ferris et al., (1998) demonstrated that runners tend to adjust leg stiffness when running on compliant surfaces in order to maintain the center of mass displacement on each stride. This autonomous process increases contact time and feasibly would decrease performance in jumping tasks on unstable surfaces. Thus, understanding how the lower limbs react under duress and fatigue may shed some light on the importance of instability training to enhance performance during jumping and cutting maneuvers.

Postural and core musculature also contribute to the success of efficient athletic movement. Throughout the literature there are a variety of descriptions or definitions for the core musculature. Behm et al., (2010) suggested that the anatomical core consisted of the axial skeleton (which includes the pelvic girdle and shoulder girdles) and all soft tissues (i.e., articular and fibro-cartilage, ligaments, tendons, muscles, and fascia) with a proximal attachment originating on the axial skeleton, regardless of whether the soft tissue terminates on the axial or appendicular skeleton (upper and lower extremities). All of these soft tissues and axial skeleton work to move in a variety of eccentric, concentric, or isometric motions (Behm et al., 2010). The importance of the core musculature in human movement is to provide a foundation for the transfer of angular momentum to the limbs and to maintain balance through postural adjustments to keep one's center of gravity (COG) over the base of support (Strange et al., 2009). Previous research has shown that muscle actions during athletic performance on unstable surfaces increase electromyographic (EMG) activity in limb and trunk muscles when being compared to stable surfaces (Anderson & Behm, 2005). Additionally, a review conducted by The Canadian Society for Exercise Physiology indicated that training under unstable conditions can significantly reduce force output in lower and upper body movements (Behm et al., 2010).

In addition to surface stability, cutting and jumping maneuver are also affected by fatigue. Tomazin and colleagues (2002) described fatigue as a disruption from the cortex to the contractile mechanisms of a muscle. Several studies have shown a decrement in balance following a protocol that fatigues the core musculature. A review conducted by Adlerton et al., (2003) concluded that trunk muscle and lower limb fatigue induce postural instability. Surenkok et al., (2008) established that trunk-muscle fatigue has an adverse effect on static and dynamic balance. Additionally, Parreira et al. (2013) presented an increase in postural sway immediately following a dynamic back extension task. All of the previous studies that measured balance following trunk fatigue protocols tested static and dynamics balance tasks. To our knowledge, very few or no studies have examined the effects of trunk fatigue and ballistic jumping movements.

In terms of muscle activity, it has been suggested that electromyographic (EMG) activity increases when moving either isometrically or dynamically on an unstable compared to a stable surface (Anderson & Behm 2005). However, the literature is not conclusive in this area. Several studies have shown a decrease in muscle activity when exerting force on unstable surfaces, particularly for lower limb exercises (Anderson & Behm 2005; McBride et al., 2006; McBride et al., 2010; Bressel et al., 2009; Saeterbakken & Fimland, 2013). For instance, McBride et al. (2010) showed a decrease in lower limb muscle activity when performing squats on an -unstable as compared to a stable surface. However, these studies reported the effects of surface instability during the performance of isometric and dynamic squats. To the authors' knowledge, there is no study available that investigated the influence of unstable surfaces on activity of lower limb as well as trunk muscles during maximal jumping and cutting tasks. During cutting (i.e., lateral jumps) and jumping, it seems that EMG activity in lower limb muscles is preprogrammed during the preactivation phase (Dyhre-Poulsen et al., 1991; Avela et al., 1996) and affected by stretching loads (Avela et al. 1996, Komi & Gollhofer 1997, Fleischmann et al., 2010; Hoffrén et al., 2011) during the braking phase. In fact, it has been shown that muscle preactivation is related to the appearance and magnitude of spinal stretch reflexes during ground contact of drop jumps (Avela et al. 1996). In this regard, jumping and landing on unstable/foam surfaces may dampen the impact at ground contact which could reduce both, muscle preactivation and reflex activity.

To the authors' knowledge, there is no study available that investigated the influence of fatigue and the interaction of unstable surfaces on activity of lower limb as well as trunk muscles during jumping and cutting maneuvers. Therefore, the objectives of this study were to investigate the effects of back extensor fatigue on *(a)* performance during jumping on stable and unstable surfaces and *(b)* activity of lower limb and trunk muscles. The literature suggests that the core musculature plays a major role in controlling posture and balance. Subsequently, performance in jumping and cutting maneuvers relies heavy on maintenance of balance. Furthermore, performance decrements are often observed on unstable surfaces. Additionally, fatigue has a detrimental effect on balance, muscle activity and force output. Thus, we hypothesized that performance measures decrease during jumping particularly on unstable surface following a fatigue protocol of the back extensors. Further, lower peak leg and trunk muscle activities are expected in the fatigued as compared to the non-fatigued condition.

Methods

Participants

With reference to the study of Wadden et al. (2012), an a priori power analysis (Faul et al. 2007) with an assumed Type I error of 0.05 and a Type II error rate of 0.20 (80% statistical power) was calculated for measures of isometric squat performance and revealed that 14 participants would be sufficient for finding medium surface x test interaction effects. None of the male subjects (Age 22.6 ± 5.4, Body mass [kg] 79.1 ± 9.6, height [cm] 178.9 ± 7.6, Body Mass Index [kg/m²] 24.7 ± 2.7, Sports activity level [h/wk] 10.3 ± 4.0) had an history of musculoskeletal, neurological, or orthopedic disorder that might have affected their ability to execute the experimental protocol. All participants were classified as physically active according to the Freiburg questionnaire for everyday and sports-related activities (Frey et al., 1999) and all had at least 5 years of experience participating in jumping sports (volleyball, basketball, soccer). All subjects read and signed a consent form prior to experimentation. Memorial University of Newfoundland's Human Investigation Committee provided ethical approval for the study.

Field Code Changed

Experimental procedure

A single-group, repeated-measures design was used to assess measures of jumping performance on stable and unstable surfaces as well as lower limb and trunk muscle EMG activity pre and post fatigue. Following a standardized warm-up protocol for the lower limbs (2 x 10 lateral shuffles with 30 seconds between trials), the maximal lateral jumping distance was assessed. To assess the lateral jumping distance, subjects jumped off the non-dominant leg and immediately upon landing with the dominant leg, jumped laterally back to the starting position with the dominant leg (Coren, 1993). Subjects were instructed not to cross their legs at any point and could not stop at the distal landing point. Following the warm-up, maximal vertical countermovement jumps and submaximal lateral jumps were performed under stable and unstable (i.e., AIREX® balance pad on top of the AMTI force plate) conditions in a randomized order. Following the initial CMJs and LJs, the modified Biering-Sørensen test (Pitcher et al., 2007) was conducted three times until failure. Between the fatigue trials, a 30 s rest period was provided. Immediately after the fatigue protocol, participants performed the same sequence of jump tests as during the non-fatigued condition.

Assessment of countermovement jump and lateral jump performance

Participants performed maximal vertical countermovement jumps while standing on a three-dimensional force plate (AMTI, Watertown, MA, USA). In accordance with the Fleischmann et al., (2010) protocol, lateral jumps were performed at 85% of the initially determined maximal lateral jumping distance. Starting from a stance position with two feet on the ground, subjects jumped laterally with their non-dominant leg onto the force plate, landing on their dominant leg and as quickly as possible returned back to the starting position. Landing on the force plate was performed one-legged (i.e., dominant leg), forefoot first, and oriented on a mark fixed perpendicular to the direction of motion onto the force plate. The subjects were instructed to jump back from the force plate to their starting position as fast as possible, omitting trunk rotations, and to keep the jumping technique similar throughout the measurements. These requirements were controlled visually using force plate data.

All LJs and CMJs on stable and unstable conditions were performed on a force plate (Three dimensional force plate, AMTI, Watertown, MA, USA), which measures vertical ground reaction force (GRF). Synchronization of GRF and EMG data was achieved by analog-to-digital conversion using a trigger connecting the Biopac EMG hardware (Biopac Systems Inc. DA 100 and analog to digital converter MPI00WSW) to the force plate A/D board, with a sampling frequency of 2000 Hz. Vertical jump height and takeoff velocity was analyzed for jumping and landing tasks and normalized to body mass. Regarding lateral jumps, GRF was used to determine contact time. In terms of the CMJ task on stable and unstable surfaces, the force time curve was used to detect braking phase, push off phase and onset of force to take-off.

Assessment of muscle activity during countermovement jumps and lateral jumps

Circular bipolar surface electrodes (Kendall 133 Foam electrodes with conductive adhesive hydrogel, Covidien, Mansfield, MA, USA), 13 mm, center-to-center distance: 25 mm) were used to measure EMG activities of 4 leg muscles (vastus medialis [VM], biceps femoris [BF], gastrocnemius medialis [GM], tibialis anterior [TA]) and 4 trunk muscles (external oblique, internal oblique, erector spinae low (L3 vertebrae), erector spinae upper (T6 vertebrae). The leg and trunk muscles were analyzed on the dominant side using the lateral preference inventory (Coren, 1993). Electrodes were positioned on the muscle bellies according to the European recommendations for surface electromyography (Hermens et al. 1999). The longitudinal axes of the electrodes were in line with the direction of the underlying muscle fibers. Inter-electrode resistance was kept below 5 kΩ by shaving, slightly roughening, degreasing and disinfecting the skin using alcohol wipes. The EMG signals were amplified and recorded with lead cables

(Biopac Systems Inc. DA 100 and analog to digital converter MPI00WSW) to a computer at a sampling frequency of 2,000 Hz. After removal of heart muscle electrical activity artifacts from the trunk muscle signals by combining adaptive filter methods with a pattern recognition mode (Konrad 2005), the filtered (10-500 Hz bandwidth), full-wave route mean squared (RMS) signals of the investigated leg and trunk muscles were triggered on the instant of ground contact and averaged over 2 countermovement jumps and 2 lateral jumps trials respectively. To find out differences in muscle activity between test conditions in counter jump performance, mean average voltage (MAV; defined as iEMG normalized relative to the integration time) was calculated for the breaking phase, push off phase, and onset-of-force to take off (Hoffrén et al. 2011). Integrated EMG (iEMG) parameters of lateral jumps were analyzed between -30 – 0 ms, 0-30 ms, 30 – 60 ms, 60-90 ms, and -90-120 ms epochs. All testing was performed in one session and electrodes were not removed, therefore normalization of iEMG and MAV was not necessary (Fleischmann et al. 2010).

Fatigue protocol

The posture adopted for the test was a variation of the Biering-Sørensen test (Pitcher et al., 2007). The Beiring-Sørensen test was originally described by the authors as having subjects lay prone on an examination table and maintain an unsupported trunk (from the superior border of the iliac crest) horizontally until they could no longer hold a horizontal position or for a maximum of 240 seconds. The buttocks and legs were fixed to the table with three, three-inch canvas straps. Any variations from the described methods are known as modified Beiring-Sørensen tests. Our tests differ from the original by not stopping the test at the recommended default of 240 seconds. All protocols were held to exhaustion (deviation from the horizontal plane). Subjects lay prone on a padded examination table, with the trunk of the body extended off the edge of the table at the level of the anterior superior iliac spine of the pelvis. The lower legs, thighs and mid-buttocks region were restrained from motion using wide straps attached to the examination table.

Statistical analyses

Data are presented as group mean values ± standard error (SE). After normal distribution was examined (i.e., Kolmogorov-Smirnov-Test), a separate 2 (surface: stable, unstable) × 2 (tests: pre, post fatigue) analysis of variance (ANOVA) with test as repeated within-subject factor was used to analyze performance and muscle activation parameters. Post-hoc tests (paired t tests) were conducted to identify the comparisons that were statistically significant. The classification of effect sizes (f) was determined by calculating partial eta-squared (η_p^2). The effect size is a measure of the effectiveness of a treatment and it helps to determine whether a statistically significant difference is a difference of practical concern. Effect sizes can be classified as small ($0.00 \leq f \leq 0.24$), medium ($0.25 \leq f \leq 0.39$), and large ($f \geq 0.40$) (Cohen 1988). The significance level was set at $p< 0.05$. Tendencies towards significance were denoted as $0.051 \leq p < 0.1$. All analyses were performed using Statistical Package for Social Sciences (SPSS) version 22.0.

Field Code Changed

Results

Countermovement jump performance

Results for CMJ and lateral jumping performance are presented in Table 2. CMJ analyses did not show any statistically significant results for test × condition**.** However, a main effect of test did significantly decrease jump height by 3.6% (p = 0.007, f = 0.57, figure 1).

Muscle activity during CMJ

EMG analysis showed that fatigue down regulated muscle activation 43.4 % and 15.7% in both the biceps femoris (p = 0.008, f = 0.58, figure 2a) and the gastrocnemius (p = 0.045, f = 0.422, figure 2b) respectively during the push-off phase with the gastrocnemius between the NFS and NFU.

A significant (p = .04, f = .438) main effect of surface was observed in the gastrocnemius. The non-fatigued unstable CMJ gastrocnemius EMG activity was 13.8% lower than the non-fatigued stable condition during the onset of force phase.

Later jump performance

There were no significant differences observed in lateral jump performance from main effect of test or main effect of surface.

Muscle activity during lateral jumps

-Fatigue down regulated muscle EMG activity by 28.3-% in the tibialis anterior (p = .05, f = .405, figure 4b) during phase 1 (-30 – 0 ms) of the lateral jump. Additionally, co-activation increased by 34.0–% (p = .03, f = .526, figure 4c) when comparing stable to unstable lateral jumps. Although non-significant (p = .06, f = .385; p = .08, f = .353), a notable trend of test x condition interaction was observed in phase 1 and phase 3 of the lateral jump for the BF. The BF increased muscle activity during phase 1 by 39.9% and 40.0% during phase 3, when comparing non-fatigued unstable to fatigued unstable (figure 4a). Additionally, the tibialis anterior muscle activity was significantly lower (28.3%, p = .05) in phase 1 (-30 – 0 ms) following the fatiguing protocol (Figure 4b).

A significant increase of 34.0% (p = .03, f = .526, figure 3b) indicating a main effect for stability was found in the co-activation of the anterior tibialis/gastrocnemius. Additionally, a significant (p = .01, f = .548) test x condition interaction was observed in the IO. During phase 5 of the fatigued unstable lateral jump the IO was 49.1 % higher compared to the fatigued stable condition. The BF showed a near significant (p = .06, .385) increased in muscle activity by 31.6% in the FU compared to the FS condition.

Discussion

Jumping performance affected by fatigue

The purpose of the present study was to investigate the relationship between fatiguing muscles of the posterior chain of both the trunk and leg muscles and compare muscle activity/performance on stable and unstable surface. The most unique finding observed was the significant decrease in the CMJ height following fatigue. Stabilization of the trunk plays a significant role in the transfer of forces and angular momentum. In multi-joint movements, the activation patterns typically follow a proximal to distal sequence, especially in locomotion and jumping movements (Kopper et al., 2012). Stabilizing and correcting the trunk's posture allows for one to position their center of gravity over their base of support. When the COG is not over the base of support, there is a loss of balance and movement may not be efficiently transferred in the desired plane motion.

Research in the area of jumping mechanics demonstrated that in order to maximize jump height, that all forces must be transferred into the vertical plane (Bobbert et al., 2011). When postural muscles become fatigued, there is a tendency for an increased postural sway and lack of postural control. These mechanisms may cause a subject to displace their center of gravity in a sub-optimal position and displace ground reaction forces in a more horizontal direction, thus not reaching maximal jump height. Kean and Behm (2006) discovered that CMJ height increased following a 6-week fixed foot balance training regime and speculated that a balance training-induced decrease in postural sway may have resulted in reaction forces being applied in a more vertical direction. This illustrates the importance of balance in jumping and supports our findings that a decrement in balance could cause a decrease in CMJ height. A study by Surenkok and colleagues (2008) used an isokinetic machine to induce trunk fatigue and tested for lactate accumulation as well as dynamic balance test and found a significant positive correlation with lactate buildup and a decrease in dynamic balance. This also agrees with our findings that fatiguing the posterior chain does affect a person's ability to correct posture and maintain balance. Additionally, a disparity amongst control and musculoskeletal properties leads to an unbalanced increase in segment angular velocities, causing the concentric velocity of some muscles to be disproportionally high and the total work produced to be unnecessarily small (Bobbert et al., 2011), basically, making a skilled jumper appear to be un-coordinated and unfamiliar with the movement.

Effects of fatigue on muscle activity

The other main finding from this study was that, a decrease in lower limb muscle activation was observed with trunk fatigue. These findings comply with the literature. Many studies have shown that fatiguing activities have a down regulating effect on EMG activity in the affected or target muscle (Oliver et al., 2008; Chan et al., 2014; Gutierrez et al., 2011). There are a number of possible mechanisms that can provide insight into the down regulation of muscle activation following back extensors fatigue. Firstly, the modified Biering-Sørensen test used in the fatiguing protocol was designed to fatigue the lower back extensors muscles. However, the method of lying prone on the table with the anterior superior portion of the iliac crest protruding over the edge, forced the participants to contract more of the muscles within the posterior chain than simply the lower back extensors. With straps, padding and support over the hamstrings and ankle, the hamstrings as well as the anterior tibialis were contracted to maintain the desired position (no deviation of the trunk from the horizontal plane). These limb contractions could potentially explain the decreased activation of both the GAS and TA following the fatigue protocol. Secondly, the build-up of lactate and other metabolites due to fatigue could also create fatiguing effects in neighboring muscles by shuttling metabolites to non-active tissue (we did not test for lactate in this investigation). This could also explain why muscles of the lower limbs were affected by a protocol that was meant to primarily fatigue the lower back extensors. Finally, the test included three repetitions to exhaustion, perhaps causing general fatigue in the central nervous system. While there are conflicting studies in the literature, a number of studies have documented fatigue of non-target or non-localized muscles. Rattey and colleagues (2005) examined the effects of cross-over fatigue by isometrically fatiguing the dominant leg and proceeded to measure the EMG activity. Following the fatiguing protocol, they observed a decrease in voluntary action as well as a decrease in iEMG activity. They cited that centrally mediated mechanisms may have been the reason for changes in the non-exercised leg. Another study that looked at cross-over fatigue discovered a post-fatigue decline in CNS excitability (Post et al., 2008). Thus, explaining a possible physiological mechanism that contributes to fatigue in a non-exercise muscle. Furthermore, another study that investigated the effects of fatiguing the hand flexors using an isometric handgrip contraction showed a temporary decrease in EMG activity in the non-exercised plantar flexor muscles (Kennedy et al., 2012). This article also indicated that the detriment in performance in the plantar flexors affected by systematic central fatigue. These articles support our findings that fatigue in one area muscle group can negatively affect performance and contractile properties in another. Unfortunately, we did not use any neuromuscular tests during this investigation to verify changes in contractile properties.

Effects of surface instability on muscle activity and performance

One main finding observed in this study was that back extensor fatigue equally affected jump performance and trunk and lower limb muscle activities while performing on stable and unstable surfaces. In addition, irrespective of muscle fatigue, there was an absence of change in performance (CMJ and LJ) and muscle activation with experienced jump-trained individuals in response to a moderately unstable foam pad. This finding applied to both jumping activities: CMJ and LJ. The findings of this investigation were similar to previous research. Prior research has shown that athletes with greater training experience will be affected to a lesser degree by an unstable surface (Wahl & Behm, 2008). Additionally, a review article concluded that athletes have better dynamic and static balance compared to the non-athletic population (Hrysomallis, 2011). Hrysomallis (2011) suggests that the improvement in performance could be a greater proprioceptive sense or simply that athletes become more skilled at focusing and attending to important sensory cues with training. Another study reported no significant stability-related differences in activation of the trunk musculature during bilateral dynamic chest press and shoulder press (Behm et al., 2005). Furthermore, Anderson and Behm (2004) were unable to show a distinction in muscle activation between stable and unstable bench press in resistance trained men. Some investigations have found little or no difference in muscle activity when comparing movement on stable to unstable surfaces for several reasons. Firstly, a moderately compliant surface allows for the elastic recoil of energy and has a trampoline-like effect (Arampatzis et al., 2004). Secondly, with skilled jumpers, a moderately unstable surface may not change the jumping strategy and thus not affect performance (Ferris & Farley, 1997). When examining the effects of core stability training in experienced runners, Sato and Mokha (2009) could not establish a significant link between core training over six weeks and any improvement in lower limb stability or ground reaction forces. They suggested that the mechanics of running were not affected with experienced runners when core musculature conditioning is improved. However, there is not unanimity in the literature. For instance, the throwing velocity of female handball players following a 6-week core stabilizing regime, was reported to significantly increase (Saterbakken et al., 2011) suggesting that a high level of core stability and strength may be required for generating force in multi-segmental movements. Although Saterbakken and colleagues were able to show differences in performance from training the trunk, their study was a training study that used upper body movements, whereas our investigation only consisted of one session and investigate lower body movements. However, other studies have been able to show a change in muscle activity due to unstable surfaces. For instance, a study that investigated muscle activity when performing isometric squats on unstable surfaces showed 37.3% and 34.4% decreases in the vastus lateralis and vastus medialis muscles (McBride et al., 2006). Another recent study that examined the effects of instability and drop jumping performance displayed a decrease in muscle activity in the gastrocnemius, vastus medialis, and biceps femoris when jumping on an unstable foam pad (Prieske~~,~~ et al., 2013). Conversely, Anderson and Behm (2005) and Bressel et al. (2009) observed increased muscle activity during the performance of dynamic lower body exercises on unstable devices (e.g., squats). On the other hand, in agreement with the reported studies investigating muscle activity with instability, it appears reasonable to argue that the studies of Anderson and Behm (2005) and Bressel et al. (2009) could have had methodologically limitations by using the same absolute weight for the stable and unstable surface condition. This argument is supported by findings from McBride et al. (2010), who reported similar activity for the spinal erector muscle during dynamic squats on stable and unstable surfaces when the same relative load was used.

Test x surface

There was no significant interaction between test and surface. The reason for the missing interaction effect was that there was no main effect for surface amongst the group.

Limitations

The level of stability was low to moderate and may not have provided sufficient perturbation to affect the experienced jumpers. However, the unstable foam pad used is frequently used in training and rehabilitation.

Conclusions

The main finding of this study was that back extensor fatigue can significantly impact jumping performance. By improving back extensor endurance and strength, individuals may be able to increase/maintain jump height in their respective competitions. As previously mentioned, due to the experienced jumping group used, there was no impact of surface stability on jumping performance. The robust fatiguing protocol used fatigued did effect performance in counter movement jump height. To investigate the phenomenon of trunk fatigue and surface instability further, a more compliant instability device could be used. Alternatively, using a non-experienced group or more robust cutting maneuver (i.e. T-Test agility run) might be used.

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Appendix – figures and tables

Table 1 - Means and standard errors of the EMG activity, measured in millivolts, for upper body muscles in the lateral jumping task. Lateral jump is broken down into 5 phases: phases 1 (-30-0ms) phase 2 (0-30ms), phase 3 (30-60ms), phase 4 (60-90ms), and phase 5 (90-120ms). NF = non-fatigue and F = fatigued

	Phases of jump (30 ms epochs)

Lateral jump Upper Body	1		2		3		4		5	
Stable	*M*	*SE*	*M*	*SE*	*M*	*SE*	*M*	*SE*	*M*	*SE*
Internal oblique (NF)	0.2122	0.038	0.2140	0.046	0.2115	0.031	0.1789	0.028	0.1555	0.028
Internal oblique (F)	0.1760	0.029	0.1869	0.029	0.1694	0.022	0.1704	0.031	0.1063	0.012
External oblique (NF)	0.0262	0.004	0.0246	0.003	0.0272	0.004	0.0299	0.004	0.0306	0.004
External oblique (F)	0.0238	0.004	0.0226	0.004	0.0242	0.004	0.0229	0.003	0.0261	0.005
Erector spinae low (NF)	0.0969	0.016	0.1106	0.016	0.1116	0.012	0.1290	0.022	0.1220	0.021
Erector spinae low (F)	0.1112	0.038	0.1126	0.033	0.1455	0.049	0.1122	0.021	0.1215	0.035
Erector spinae high (NF)	0.0558	0.011	0.0973	0.015	0.1034	0.021	0.1153	0.023	0.1121	0.015
Erector spinae high (F)	0.0768	0.016	0.1005	0.014	0.1003	0.011	0.0926	0.017	0.1116	0.014
Unstable										
Internal oblique (NF)	0.1512	0.032	0.1815	0.033	0.2184	0.038	0.2002	0.040	0.1202	0.014
Internal oblique (F)	0.2119	0.042	0.2250	0.043	0.2486	0.044	0.2085	0.039	0.1867	0.036
External oblique (NF)	0.0267	0.005	0.0225	0.003	0.0238	0.003	0.0249	0.003	0.0321	0.005
External oblique (F)	0.0290	0.005	0.0244	0.004	0.0265	0.003	0.0278	0.003	0.0294	0.003
Erector spinae low (NF)	0.0685	0.018	0.0697	0.015	0.0886	0.016	0.0846	0.009	0.0836	0.017
Erector spinae low (F)	0.0633	0.017	0.0803	0.024	0.1036	0.017	0.1210	0.021	0.1126	0.016
Erector spinae high (NF)	0.0479	0.009	0.0702	0.015	0.0845	0.014	0.1261	0.024	0.1089	0.018
Erector spinae high (F)	0.0738	0.013	0.1266	0.026	0.1202	0.014	0.1268	0.016	0.1180	0.016

Table 1b - Means and standard errors of the EMG activity, measured in millivolts, for the lower body muscles in the lateral jumping task. Lateral jump is broken down into 5 phases: phases 1 (-30-0ms) phase 2 (0-30ms), phase 3(30-60ms), phase 4 (60-90ms), and phase 5 (90-120ms). NF = non-fatigue and F = fatigued

Table 1b	**Phases of jump (30 ms epochs)**									
Lateral jump lower body	1		2		3		4		5	
Stable	*M*	*SE*	*M*	*SE*	*M*	*SE*	*M*	*SE*	*M*	*SE*
Vastus medialis (NF)	0.1314	0.029	0.1673	0.032	0.2195	0.042	0.2281	0.042	0.2775	0.066
Vastus medialis (F)	0.1221	0.025	0.2005	0.068	0.2278	0.079	0.2128	0.058	0.1934	0.029
Biceps Femoris (NF)	0.0805	0.014	0.0936	0.015	0.0871	0.013	0.0894	0.021	0.0834	0.014
Biceps Femoris (F)	0.0608	0.012	0.0723	0.013	0.0746	0.014	0.0814	0.017	0.0878	0.026
Tibialis Anterior (NF)	0.2551	0.024	0.2869	0.054	0.2626	0.030	0.2596	0.039	0.3592	0.065
Tibialis Anterior (F)	0.1755	0.027	0.2165	0.050	0.2443	0.051	0.2673	0.058	0.3156	0.084
Gastrocnemius (NF)	0.2519	0.043	0.2303	0.041	0.2093	0.036	0.1608	0.034	0.1598	0.036
Gastrocnemius (F)	0.2134	0.046	0.2236	0.047	0.1652	0.040	0.0988	0.018	0.0910	0.017
Unstable										
Vastus medialis (NF)	0.1336	0.041	0.1501	0.051	0.1892	0.050	0.2284	0.062	0.2734	0.056
Vastus medialis (F)	0.1969	0.087	0.2210	0.084	0.2730	0.099	0.3290	0.127	0.3599	0.115
Biceps Femoris (NF)	0.0678	0.012	0.0861	0.014	0.0781	0.014	0.0703	0.012	0.0911	0.013
Biceps femoris (F)	0.0948	0.018	0.1059	0.018	0.1090	0.016	0.0844	0.014	0.0794	0.014
Tibialis Anterior (NF)	0.2204	0.047	0.2648	0.035	0.3489	0.061	0.3465	0.048	0.3656	0.065
Tibialis Anterior (F)	0.1953	0.033	0.2648	0.037	0.3120	0.063	0.3260	0.084	0.3041	0.059
Gastrocnemius (NF)	0.2290	0.035	0.1956	0.027	0.1664	0.028	0.1368	0.032	0.1223	0.035
Gastrocnemius (F)	0.1958	0.030	0.1776	0.029	0.1558	0.029	0.1388	0.045	0.1325	0.039

Table 1c - Means and standard errors of the EMG activity, measured in millivolts, for the upper body muscles in the CMJ task. CMJ is broken down into 3 phases: Breaking phase (BP), push-off phase (PP), and the onset of force (OF). NF = non-fatigue and F = fatigued.

Table 1c	**Phases of jump**					
CMJ Upper body	Breaking phase		Push-off phase		Onset of force to take-off	
Stable	*M*	*SE*	*M*	*SE*	*M*	*SE*
Internal oblique (NF)	0.092	0.03	0.245	0.05	0.128	0.03
Internal oblique (F)	0.121	0.03	0.213	0.04	0.133	0.03
External oblique (NF)	0.029	0.01	0.058	0.01	0.036	0.01
External oblique (F)	0.039	0.02	0.061	0.02	0.062	0.03
Erector spinae low (NF)	0.192	0.04	0.199	0.03	0.148	0.03
Erector spinae low (F)	0.202	0.04	0.171	0.03	0.160	0.03
Erector spinae high (NF)	0.204	0.03	0.182	0.02	0.134	0.01
Erector spinae high (F)	0.171	0.03	0.152	0.02	0.146	0.02
Unstable						
Internal oblique (NF)	0.088	0.02	0.247	0.05	0.120	0.02
Internal oblique (F)	0.116	0.03	0.237	0.04	0.130	0.02
External oblique (NF)	0.045	0.02	0.093	0.03	0.049	0.02
External oblique (F)	0.048	0.02	0.067	0.02	0.047	0.02
Erector spinae low (NF)	0.161	0.04	0.200	0.03	0.122	0.02
Erector spinae low (F)	0.190	0.04	0.198	0.03	0.140	0.02
Erector spinae high (NF)	0.206	0.03	0.198	0.02	0.139	0.01
Erector spinae high (F)	0.204	0.02	0.190	0.02	0.152	0.02

Table 1d - Means and standard errors of the EMG activity, measured in millivolts, for the lower body muscles in the CMJ task. CMJ is broken down into 3 phases: Breaking phase (BP), push-off phase (PP), and the onset of force (OF). NF = non-fatigue and F = fatigued.

Table 1d	**Phases of jump**					
CMJ lower body	Breaking phase		Push-off phase		Onset of force to take-off	
Stable	*M*	*SE*	*M*	*SE*	*M*	*SE*
Vastus medialis (NF)	0.277	0.04	0.447	0.06	0.238	0.02
Vastus medialis (F)	0.271	0.04	0.374	0.08	0.301	0.06
Biceps Femoris (NF)	0.103	0.02	0.239	0.03	0.110	0.01
Biceps Femoris (F)	0.130	0.04	0.132	0.02	0.103	0.02
Tibialis Anterior (NF)	0.292	0.05	0.185	0.02	0.212	0.03
Tibialis Anterior (F)	0.262	0.07	0.141	0.03	0.207	0.04
Gastrocnemius (NF)	0.077	0.01	0.364	0.03	0.158	0.01
Gastrocnemius (F)	0.098	0.02	0.271	0.05	0.135	0.02
Unstable						
Vastus medialis (NF)	0.286	0.05	0.446	0.06	0.236	0.03
Vastus medialis (F)	0.289	0.05	0.425	0.06	0.276	0.05
Biceps Femoris (NF)	0.098	0.02	0.211	0.02	0.097	0.01
Biceps femoris (F)	0.083	0.02	0.182	0.03	0.110	0.01
Tibialis Anterior (NF)	0.281	0.03	0.183	0.01	0.205	0.02
Tibialis Anterior (F)	0.321	0.05	0.174	0.02	0.216	0.03
Gastrocnemius (NF)	0.094	0.01	0.348	0.02	0.139	0.01
Gastrocnemius (F)	0.094	0.01	0.345	0.04	0.149	0.01

Table 2 - Means and standard errors for CMJ and LJ performance. Results are listed pre fatigue and post fatigue protocol and present with percentage differences, significance (*p*-value) and effect size (*f*).

Table 2

Variables	**Main effect of fatigue**						
	Pre		**Post**	**(n= 14)**	**Δ (%)**	***p***	**(effect size)**
	M	*SE*	*M*	*SE*			
CMJ height (cm)	37.480	1.30	36.170	1.24	-3.622	0.007	0.570
Take-off velocity (m/s)	2.712	0.26	2.662	0.25	1.825	0.023	0.470
Lateral jump contact time (s)	0.757	0.03	0.723	0.03	-1.825		0.310
	Main effect of condition - Stable vs Unstable						
	Stable		**Unstable**	**(n= 14)**	**Δ (%)**	***p***	**(effect size)**
	M	*SE*	*M*	*SE*			
CMJ height (cm)	37.130	1.34	36.530	1.27	-1.642	0.816	0.040
Take-off velocity (m/s)	2.683	0.24	2.648	0.24	-1.318	0.023	0.470
Lateral jump contact time (s)	0.697	0.02	0.781	0.04	10.818	0.126	0.100

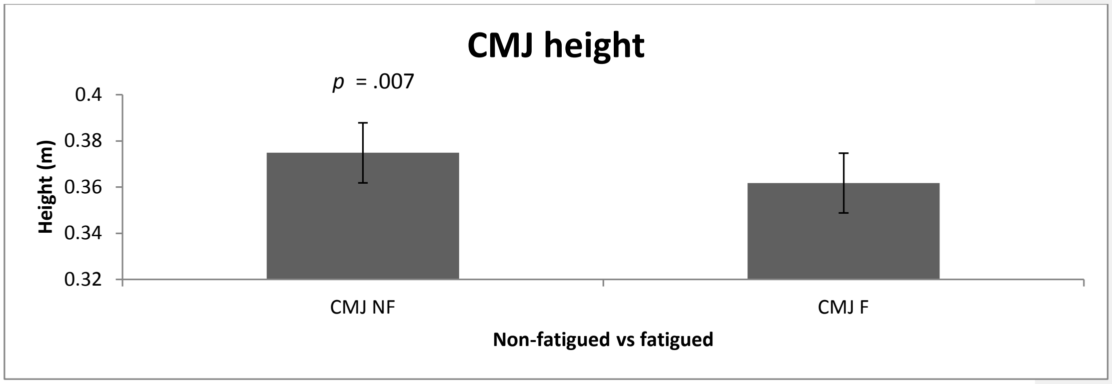

Figure 1 - Counter movement jump (CMJ) height – pre fatigue versus post fatigue. Error bars represent standard error and significant decrease post fatigue indicated by *p*=.007. X-axis represents CMJ NF (non-fatigued) and CMJ F (fatigued). Y-axis represents jump height in meters.

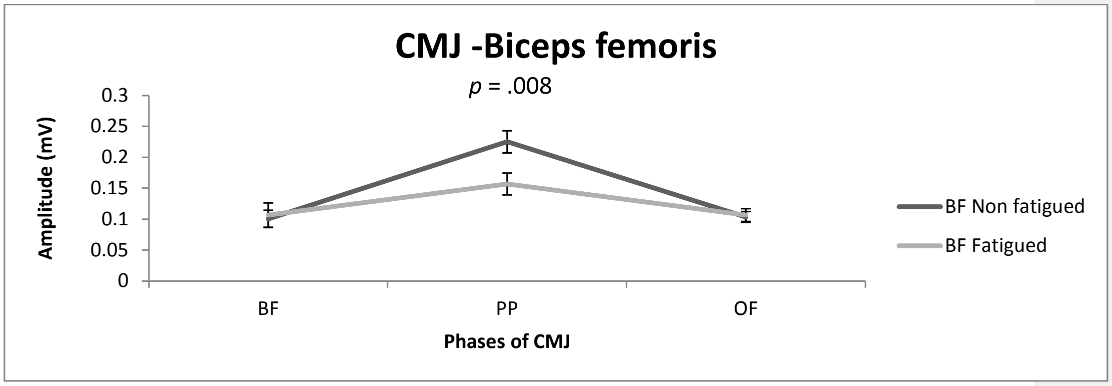

Figure 2a - Effect of fatigue on EMG activity of biceps femoris in CMJ task. Significant decrease in muscle activation observed (*p* =.008) in phase 2 (push-off phase). Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1(BP), phase 2 (PP), and phase 3 (OF).

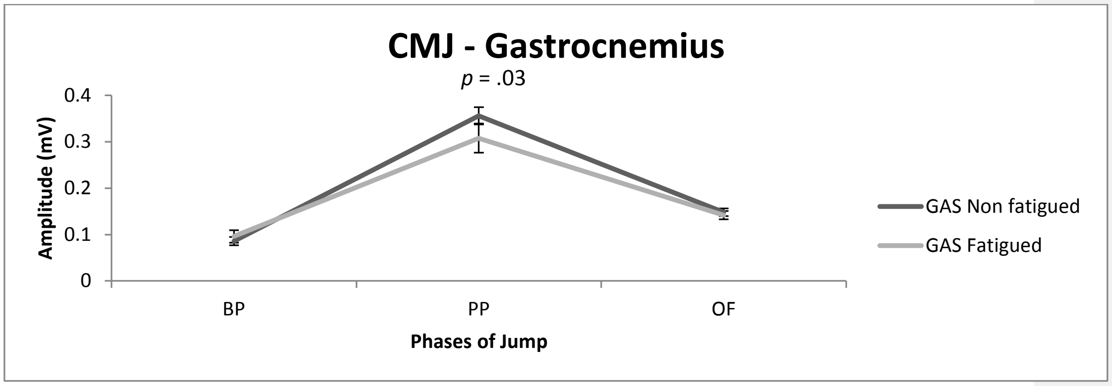

Figure 2b - Effect of fatigue on EMG activity of gastrocnemius in CMJ task. Significant decrease in muscle activation observed (*p* =.03) in phase 2 (push-off phase) following the fatiguing protocol. Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1(BP), phase 2 (PP), and phase 3 (OF).

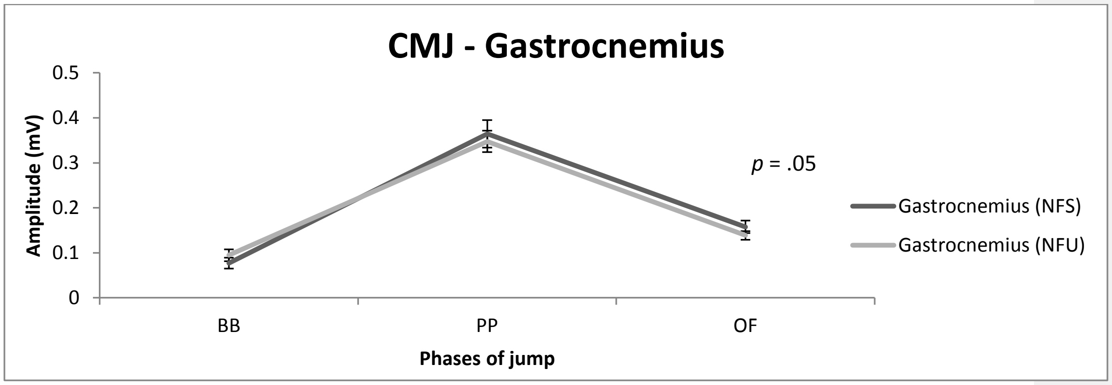

Figure 2c - Effect of test x condition interaction of gastrocnemius in CMJ task. Significant decrease in muscle activation (*p* = .05) was observed in the non-fatigued unstable condition compared to the non-fatigued stable condition. Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1(BP), phase 2 (PP), and phase 3 (OF).

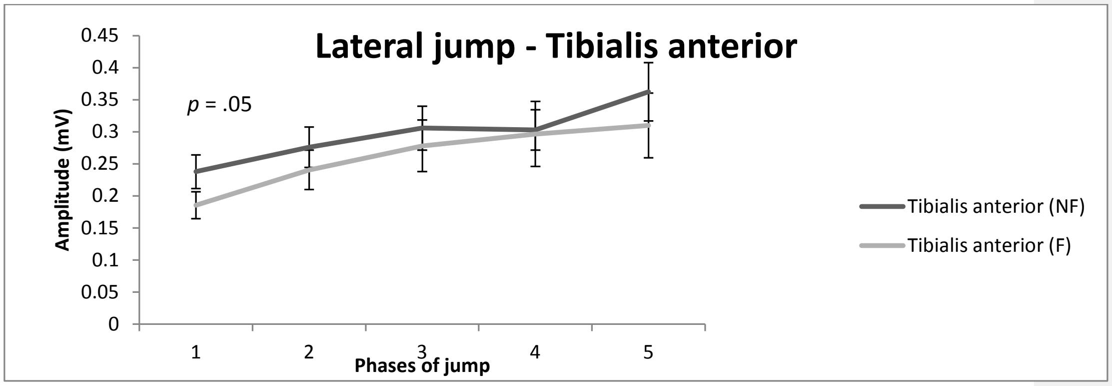

Figure 3a - Effect of fatigue on EMG activity of tibialis anterior (TA) in LJ task. Significant decrease in muscle activation observed (*p* =.03) in phase 1 (-30-0ms) following the fatiguing protocol. Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1 (-30-0ms), phase 2 (0-30ms), phase 3 (30-60ms), phase 4 (60-90ms), phase 5 (90-120ms).

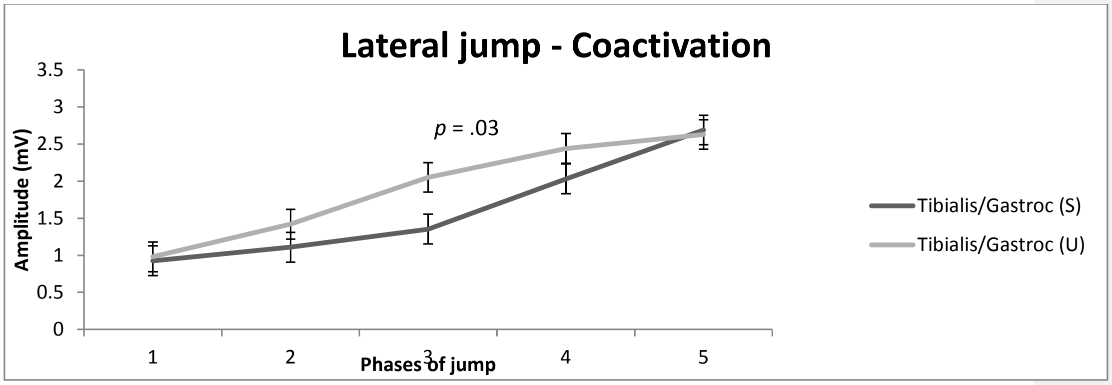

Figure 3b - Effect of surface stability on EMG activity of coactivation between tibialis anterior (TA) and gastrocnemius (GAS) in LJ task. Significant increase in muscle activation observed (*p* =.03) in phase 3 (-30-0ms) under the unstable condition. Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1 (-30-0ms), phase 2 (0-30ms), phase 3 (30-60ms), phase 4 (60-90ms), phase 5 (90-120ms).

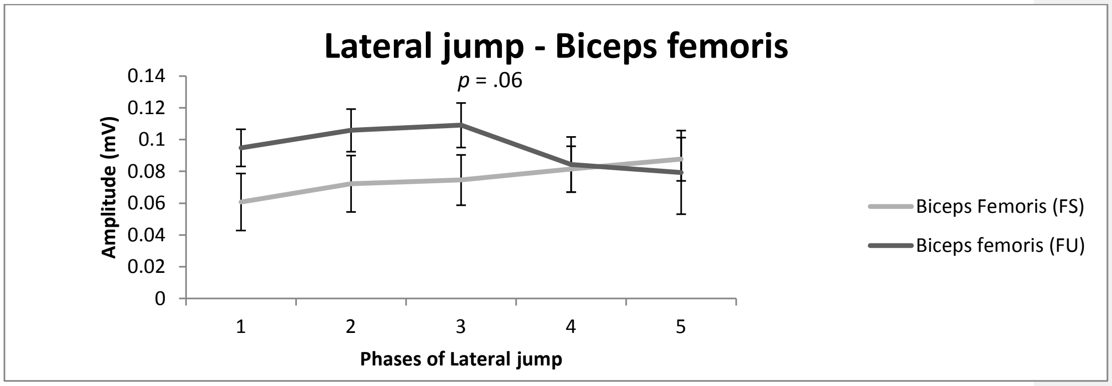

Figure 3c - Effect of test x condition interaction on EMG activity of biceps femoris (BF) in LJ task. Significant increase in muscle activation observed (*p* =.03) in phase 3 (-30-0ms) under the fatigued unstable condition compared to the fatigue stable condition. Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1 (-30-0ms), phase 2 (0-30ms), phase 3 (30-60ms), phase 4 (60-90ms), phase 5 (90-120ms).

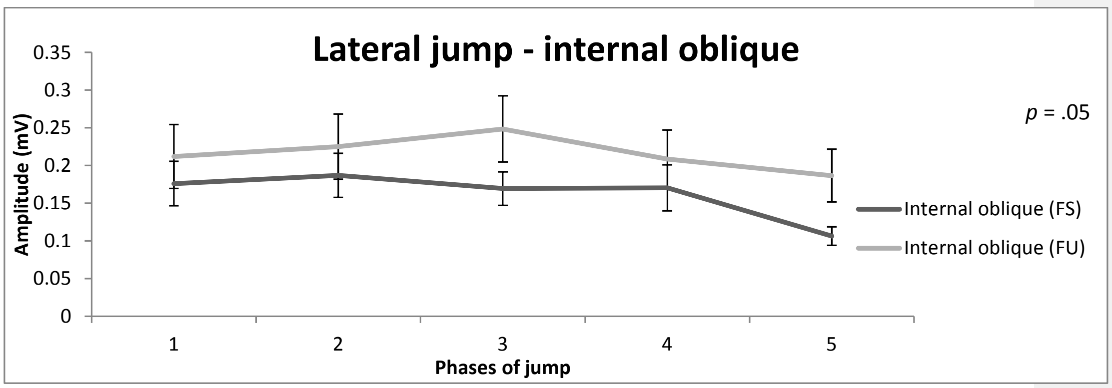

Figure 3d - Effect of test x condition interaction on EMG activity of the internal obliques (IO) in LJ task for the Fatigued stable (FS) and Fatigued unstable (FU) conditions. Significant increase in muscle activation observed (*p* =.05) in phase 5 (90-120 ms) under the fatigued unstable condition compared to the fatigue stable condition. Y-axis represents amplitude in millivolts. X-axis represents phases of jump: phase 1 (-30-0 ms), phase 2 (0-30 ms), phase 3 (30-60 ms), phase 4 (60-90 ms), phase 5 (90-120 ms).