Ankle Bracing's Effect on Lower Extremity Biomechanics During Athletic Performance Measures: An Exploratory Study

By

Zachariah J. Henderson

Supervisor: Dr. Paolo Sanzo

Committee: Dr. Carlos Zerpa and Dr. Derek Kivi

In partial fulfilment of the requirements

for the degree of Master of Science in Kinesiology

SCHOOL OF KINESIOLOGY

LAKEHEAD UNIVERSITY

THUNDER BAY, ONTARIO, CANADA

August, 2018

Abstract

Introduction: Ankle braces are commonly worn in athletic populations to prevent ankle injuries. Restriction of ankle range of motion (ROM) is considered the main mechanism by which ankle braces prevent ankle injuries. The effects of restricting normal ankle ROM on lower extremity injury, biomechanics, and athletic performance, however, remains unclear. Although the research is conflicting, increases in non-ankle lower extremity injuries when wearing ankle braces has been observed, in addition to changes in lower extremity kinematics, kinetics, vertical jump height, and agility. Decreases in vertical jump height when wearing ankle braces have previously been attributed to restriction of ankle ROM and altered lower extremity kinematics. No studies, however, have considered the effect that ankle braces may have on muscular activation of the lower extremity, specifically proximal musculature of the knee and hip joint, on athletic performance. As such, the purpose of this study was to examine the effects of softshell and semi-rigid ankle braces on muscle electromyography (EMG), kinetics, and performance during a vertical jump test and cutting task.

Methods: 42 physically active individuals (23 male, 19 female) were recruited into the study. Participants completed a Vertical Jump Test and cutting task on two separate days under three bilateral conditions: wearing no ankle braces, ASO® EVO® (AE) softshell ankle braces, or Active Ankle T1™ (T1) semi-rigid ankle braces. Vertical jump height, and mean EMG activity of the peroneus longus (PL), lateral gastrocnemius (LG), biceps femoris (BF), rectus femoris (RF), and gluteus medius (GM) muscles was collected during the landing and takeoff phase of the Vertical Jump Test. Similarly, time to compete the cutting task, as well as mean EMG activity of the peroneus longus (PL), lateral gastrocnemius (LG), biceps femoris (BF), rectus femoris (RF), and gluteus medius (GM) muscles was collected during the deceleration and propulsive phase of the cutting task. Ground reaction forces and impulse were also collected during each task and phase. Due to missing data, repeated measures one-way ANOVAs were conducted to compare the independent variable (brace condition) on each dependent variable. The alpha level was set at $p < .05$ for the performance measures. After a Bonferroni adjustment, the alpha level was set at $p < .01$, $p < .01$, and $p < .017$ for EMG, GRF, and impulse variables, respectively.

Results: There was a significant decrease in vertical jump height when wearing ankle braces, $F(2, 80) = 15.796$, $p < .001$, $\eta p^2 = .283$. Bonferroni pairwise comparisons analysis revealed a significant decrease in vertical jump height when wearing the AE (2.09 (95% CI, 0.9 to 3.28) cm, $p < .001$) and T1 (2.12 (95% CI, 1.058 to 3.193) cm, $p < .001$) ankle braces, compared to no braces. There was a significant decrease in LG mean EMG activity during takeoff when wearing ankle braces, $F(2, 68) = 5.597$, $p < .001$, $\eta p^2 = .141$. Bonferroni pairwise comparisons analysis revealed a significant decrease in LG mean EMG activity when wearing the T1 ankle braces (-7.34 (95% CI, -13.307 to -1.376) %MVC, $p = .012$), compared to no braces. There was a significant increase in peak lateral GRF during takeoff, $F(1.118, 43.585) = 39.80$, $p < .001$, $\eta p^2 = .505$. Bonferroni pairwise comparisons analysis revealed a significant increase in peak lateral GRF when wearing the AE (6.64 (95% CI, 4.066 to 9.233) %BW, $p < .001$) and T1 (6.76 (95% CI, 4.131 to 9.404) %BW, $p < .001$) ankle braces, compared to no braces.

There was a significant decrease in peak lateral GRF, $F(2, 74) = 5.746$, $p < .001$, $\eta p^2 = .134$. Bonferroni pairwise comparisons analysis revealed a significant decrease in peak lateral

GRF during landing when wearing the T1 ankle braces (-2.59 (95% CI, -4.432 to -0.764) %BW, $p < .001$), compared to no braces.

There was a significant increase in time to complete the cutting task, $F(2, 76) = 17.242$, $p < .001$, partial $\eta^2 = 0.312$. Bonferroni pairwise comparisons analysis revealed a significant increase in time to complete the cutting task when wearing the AE (0.16 (95% CI, .062 to .265) sec, $p < .001$) and T1 (0.2(95% CI, .113 to .286) sec, $p < .001$) ankle braces, compared to no braces.

Conclusion: Based on the results of this study, both softshell and semi-rigid ankle braces significantly decreased vertical jump and cutting task performance. Furthermore, ankle braces decreased EMG activity of ankle musculature and altered GRFs during vertical jumping. Further research is needed to determine how changes in EMG activity and kinetics may affect injury, as well as performance during vertical jumping and cutting. Clinicians, athletes, trainers, and any users or prescribers of ankle braces should weigh the pros and cons of prophylactically bracing the ankle, especially from a performance perspective.

Acknowledgements

As many before me would probably attest to, it is difficult to put the experience that is grad school into words. Excitement, frustration, and seemingly endless nights of work (among many other things), all dot the roller coaster that is grad school. It is even more difficult to express gratitude to all the people that helped me persist through this degree. First, I have to thank numerous times over my parents, Lesagh and Randy, for their personal (and financial) support. Without them, this degree would not have been possible. I would also like to thank Dr. Sanzo for his guidance, input, expertise, and supervision on both my Master's and undergraduate thesis. Without you, I don't think I would have developed the passion for research that I now have. Many thanks as well to my committee members, Dr. Zerpa and Dr. Kivi, for their valuable input and technical assistance on the project.

Although my name may be on the title page, a Master's degree is very much a group project; thank you to all the Duññ Bros, fellow grad students, and front desk employees for your company, friendship, advice, and assistance with procrastination. Noreen (and all the baristas at The Study): thank you for always supplying the fuel for my academic pursuits. I also have to thank the faculty gym regulars who let me informally pick their brains every morning for seven years. To everyone who was a participant in this project, thank you for volunteering your time to science.

Lastly, I would like to thank the late Brad Reeson. You taught me how to work hard, and that life is the most important thing you can train for. Without you, I don't think I would have become the person that I am today.

"What you are as a person is far more important than what you are as a basketball player."

John Wooden

"It is possible to make no mistakes and still lose. That is not weakness, that is life."

Jean-Luc Picard

#FG

Contents

List of Figures

List of Tables

List of Abbreviations

AE — ASO® EVO® softshell ankle brace

ATFL — Anterior talofibular ligament

BF — Biceps femoris muscle

CAD — Canadian dollars

CAI — Chronic ankle instability

CAIT — Cumberland Ankle Instability Tool

CFL — Calcaneofibular ligament

EMG — Electromyography

GM — Gluteus medius muscle

GRF — Ground reaction force

ICC — Intraclass correlation coefficient

LG — Lateral gastrocnemius muscle

MVC — Maximum Voluntary Contraction

mV — Millivolts

N — Newtons

PL — Peroneus longus muscle

PTFL — Posterior talofibular ligament

RF — Rectus femoris muscle

ROM — Range of motion

SEBT — Star Excursion Balance Test

T1 — Active Ankle T1™ semi-rigid ankle brace

TA — Tibialis anterior muscle

USD — United States dollars

V — Volts

Chapter One: Introduction

Overview

Ankle sprains make up the majority of musculoskeletal injuries seen in sport (Ferran & Maffulli, 2006) and represent the most frequently reported injury in the National Collegiate Athletics Association (Roos, Kerr, & Mauntel, 2016). As such, the prevention of ankle sprains and ankle injuries is of the utmost importance to an athlete, so that he/she can remain healthy and perform at his/her best. Various preventative methods are employed by athletes and healthcare providers in an attempt to prevent and treat ankle injuries, including ankle braces (Bahr, Lian, & Bahr, 1997; Kaminski et al., 2013).

While ankle braces are a commonly accepted method of preventing and/or treating ankle injuries, their use is not without controversy. Clinicians have long speculated that extended use of an ankle brace may be detrimental to muscular activation and strength about the ankle (Cordova & Ingersoll, 2003). Recent systematic reviews, however, have concluded that ankle braces can effectively prevent and treat ankle injuries in athletic populations (Leppänen, Aaltonen, Parkkari, Heinonen, & Kujala, 2014; Petersen et al., 2013). As such, the National Athletic Trainers Association (NATA) recommends that all players returning to play from an ankle sprain wear an ankle brace (Kaminski et al., 2013). Despite their widespread use and recommendation, the effect that ankle bracing has on proximal joints is not well understood. Wearing ankle braces has been noted to increase the incidence of non-ankle lower extremity injuries (McGuine, Brooks, & Hetzel, 2011; Robbins & Waked, 1998; Yang et al., 2005), as well as negatively affect athletic performance (Ambegaonkar et al., 2011; Henderson, Sanzo, & Zerpa, 2016; Parsley, Chinn, Lee, Ingersoll, & Hertel, 2013; Smith, Claiborne, & Liberi, 2016). As such, it has been suggested that further research be completed on the effects of ankle bracing on lower extremity and lumbar spine biomechanics (McGuine et al., 2011), as well as athletic performance measures (Ambegaonkar et al., 2011).

The Ankle Joint

The inferior tibiofibular joint, talocrural joint, and subtalar joint comprise what is commonly referred to as the ankle region (Figure 1). As a syndesmosis type joint, the inferior tibiofibular joint is made up primarily of fibrous tissue and many small ligaments (Brockett & Chapman, 2016). An interosseous membrane connects the tibia and fibula (Procter & Paul, 1982), maintaining the position of the tibia and fibula during movement (Brockett & Chapman, 2016). The talocrural, or ankle joint proper, is a synovial hinge joint. It allows for plantarflexion and dorsiflexion in the frontal plane (Brocket & Chapman, 2016) and is most stable in maximum dorsiflexion (Magee, 2014). The subtalar joint is the most distal joint of the ankle region and is capable of movement in three planes in non-weight bearing; dorsiflexion, abduction, eversion and plantarflexion, adduction, and inversion. In weight bearing, the subtalar joint allows for pronation and supination of the foot (Jastifer & Gustafson, 2014).

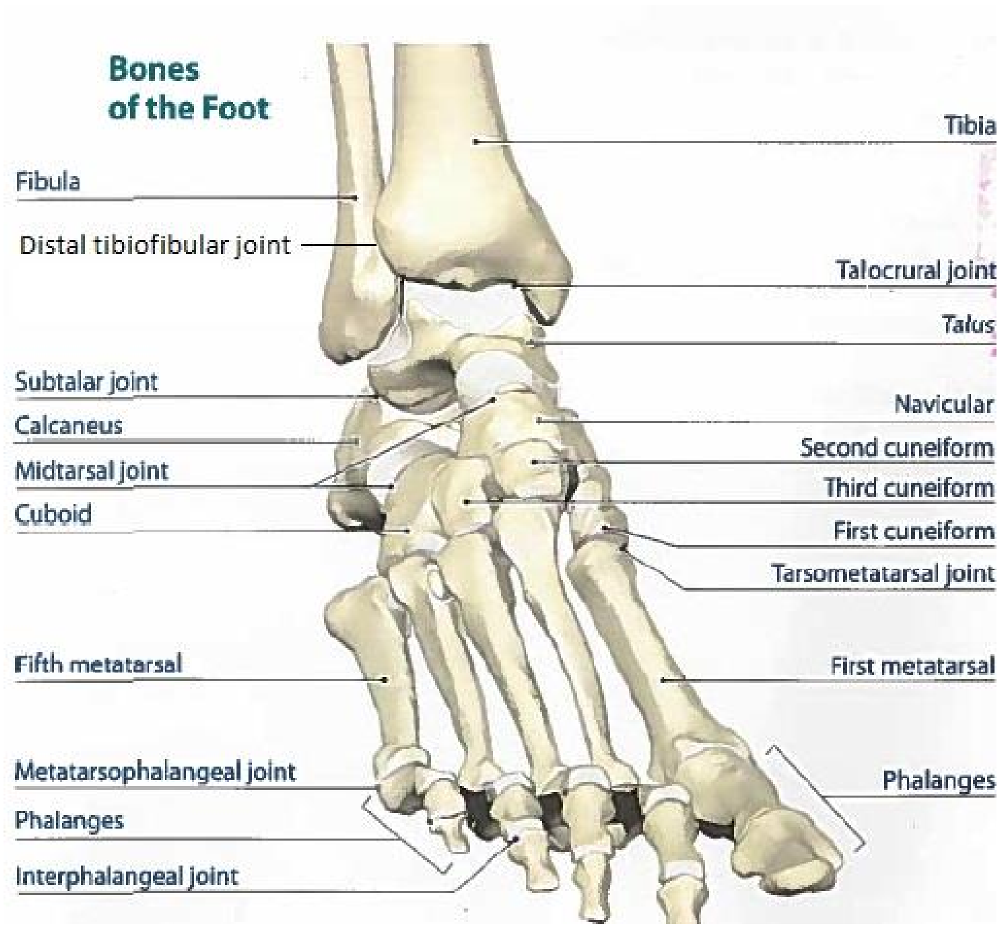


Figure 1. Joints of the ankle region. This figure displays the location of major joints of the ankle region. Adapted from “Athletic taping and bracing (3rd Ed.),” by D. G. Perrin, p. 19.

Due to the distal location of the ankle, it functions both to support and to propel the body through space (Brocket & Chapman, 2016; Magee, 2014). Additionally, it is responsible for dissipating forces that act on the body during contact with the ground or other surfaces (Donatelli, 1987; Zhang, Bates, & Dufek, 2000). As a result, the ankle is placed at a higher risk for soft tissue sprain injuries than most joints. In 2010, there were a reported 225,114 emergency room visits for acute ankle sprains in the United States (Shah, Noone, Blanchette, & Wikstrom, 2016). Ankle sprains are thought to account for at least 45% of all musculoskeletal injuries in organized sport (Ferran & Maffulli, 2006), occurring most frequently in basketball players (Roos et al., 2016). Furthermore, approximately 85% of all ankle sprains are classified as inversion type ankle sprains (Ferran & Maffulli, 2006).

Ankle sprains. Acute ankle sprains occur when one or more of the supporting ligaments of the ankle is stretched or torn (Figures 2 and 3). The most common mechanism of injury is excessive inversion of the foot while in plantarflexion, coupled with external rotation of the leg, forefoot abduction, and internal rotation of the hindfoot (Martin, Davenport, Paulseth, Wukich, & Godges, 2013). Often, this can occur when landing on an unstable surface from a jump or when making a sharp change in direction (Bahr, Karlsen, Liam, & Øvrebø, 1994; Ferran & Maffulli, 2006). The ligaments most often affected with a lateral ankle sprain include the anterior talofibular ligament (ATFL), calcaneofibular ligament (CFL), and the posterior talofibular ligament (PTFL).

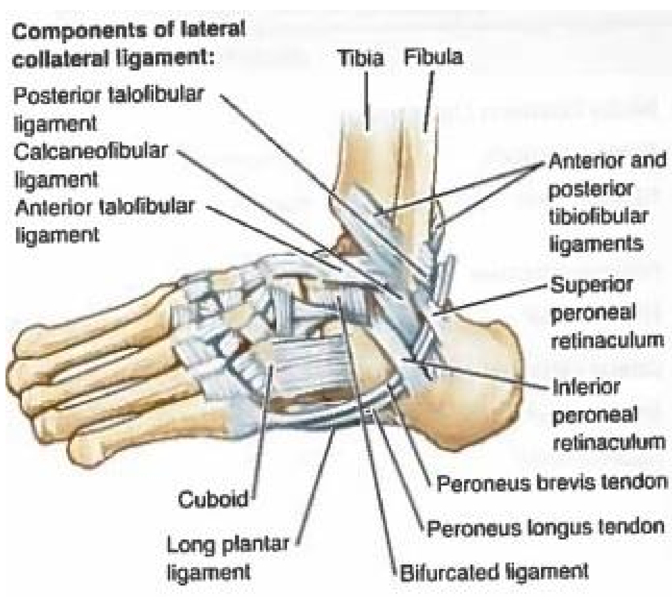


Figure 2. Lateral ankle ligaments. This figure displays the major ligaments, tendons, and bones of the lateral aspect of the foot and ankle. Adapted from “Foundations of athletic training (5th ed.),” by M. K. Anderson and G. P. Parr, p. 691.

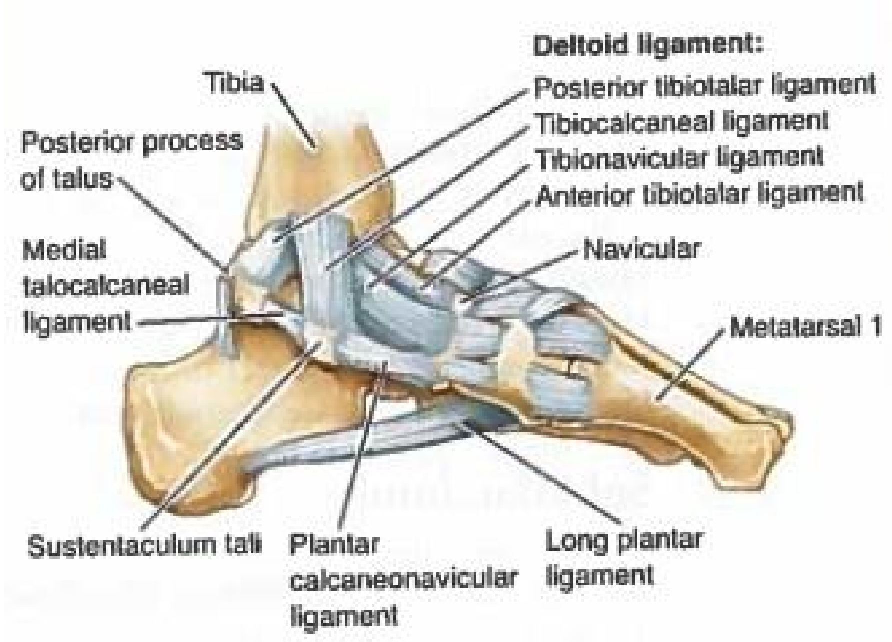


Figure 3. Medial ankle ligaments. This figure displays the major ligaments, tendons, and bones of the lateral aspect of the foot and ankle. Adapted from "Foundations of athletic training (5th ed.)," by M. K. Anderson and G. P. Parr, p. 691.

The ATFL is the weakest and most commonly affected ligament in an acute lateral ankle sprain (van den Bekerom, Oostra, Golanó, & van Dijk, 1994; Sauer, Jungfer, & Jungbluth, 1978; van Dijk, 2008). It connects the anterior and distal portion of the lateral malleolus to the body of the talus. When the ankle is in plantarflexion, the ATFL functions to limit displacement of the talus. Thus, it is also responsible for limiting inversion while the ankle is plantarflexed (van den Bekerom et al., 2008). As such, it is the primary ligament affected when the ankle is plantarflexed and inverted at the moment of injury (Holmer, Sondergaard, Konradsen, Nielson, & Jorgensen, 1994).

The CFL connects the lateral and distal portion of the lateral malleolus to the calcaneus, traversing both the talocrural and subtalar joints (Martin et al., 2013), with the fibres of the ligament running parallel to the axis of the subtalar joint (Safran, Benedetti, Bartolozzi, &

Mandelbaum, 1999). As such, it is the first ligament to be affected when the ankle is in inverted and dorsiflexed at the moment of injury (Colville, Marder, Boyle, & Zarins, 1990).

The PTFL is the strongest of the lateral ankle ligaments, connecting the posterior medial aspect of the medial malleolus to the lateral tubercle of the talus (van den Bekerom et al., 2008). Due to its anatomical properties, it is rarely injured with a lateral ankle sprain. Rather, the PTFL is usually injured when the foot is in dorsiflexion and external rotation at the moment of injury (Butler & Walsh, 2004).

While each lateral ligament has a primary mechanism of injury, multiple ligaments and structures can be affected depending on the severity of the injury. For example, in an acute ankle sprain involving plantarflexion and inversion, the ATFL is the first ligament to be stressed. As the ATFL is compromised and the inversion angle increases, the CFL can be compromised (Anderson & Parr, 2013). Furthermore, the lateral ankle ligaments are not the only structures that can be affected by the ankle sprain. If the inversion stress is great enough, the deltoid ligament on the medial aspect of the foot can be compressed resulting in injury (van Dijk, Bossuyt, & Marti, 1996). Although not as common, the deltoid ligament can also be damaged if the injury mechanism involves dorsiflexion and eversion (Anderson & Parr, 2013). Damage to the articular structures, syndesmosis, strains of the peroneus longus (PL) and brevis tendons, neuritis of the intermediate and medial dorsal cutaneous nerves, peroneal nerve, and posterior tibial nerve, avulsion fractures of the base of the fifth metatarsal, as well as the lateral and medial malleoli can all occur concurrently with an acute ankle sprain (Fallat, Grimm, & Sarraco, 1998; Nitz, Dobner, & Kersey, 1985).

Ankle sprain classification. Acute lateral ankle sprains are often classified on a three grade system of severity, as described by Dutton (2012), Magee (2014), and Reid (1992). Persons presenting with a grade one sprain generally experience tenderness over the injured ATFL with palpation. Any edema will be minimal and localized to the area of the ATFL. Partial to full weight bearing is usually possible. There is no significant tearing of the ATFL with a grade one sprain; as such, no instability will be present during ligament stress testing. Recovery from a grade one sprain can range from two to 10 days (Reid, 1992).

With a grade two acute lateral ankle sprain, the ATFL and CFL may be compromised and tender on palpation. There may be moderate edema, local to the ATFL and CFL. Persons with a grade two ankle sprain often have difficulty bearing weight on the affected limb without external support or crutches. As such, a noticeable limp will be present during gait. Furthermore, persons suffering from a grade two ankle sprain will be unable to run or hop and have difficulty with dorsiflexion and plantarflexion. A partial or full tear of the ATFL, as well as a partial tear of the CFL is possible with a grade two sprain. This results in slight instability during ligament stress testing. Recovery from a grade two sprain can range from 10 to 30 days (Reid, 1992).

With a grade three ankle sprain, the ATFL, CFL, and PTFL may be injured and tender on palpation. Significant edema will be present locally and possibly on the medial and posterior aspect of the foot. Weight bearing is not possible without significant discomfort and most people will present with little to no active range of motion (ROM) at the ankle. A full tear of at least two ligaments, as well as a partial tear of a third ligament may be present with a grade three sprain. As such, there will be defined instability during ligament stress testing. In some cases, persons with a grade three sprain may report hearing or experiencing a snapping sensation at the time of injury. Recovery from a grade three sprain can range from 30 to 90 days (Reid, 1992).

Several interventions for treating acute ankle sprains exist; however, the degree to which these interventions are supported by the literature varies. The NATA and American Physical Therapy Association recommend the application of cryotherapy to control edema, minimize secondary injury, and manage pain during the acute phase of an ankle sprain (Kaminski et al., 2013; Martin et al., 2013). Non-steroidal anti-inflammatory medications may also be used to reduce edema and pain following an ankle sprain (Kaminski et al., 2013; Mazieres, Rouanet, Velicy, Scarsi, & Reiner, 2005; Slatyer, Hensley, & Lopert, 1997). In the case of grade one and grade two ankle sprains, external support, such as a softshell ankle brace is recommended to allow for progressive weight bearing on the affected limb as soon as possible (Kaminski et al., 2013; Martin et al., 2013). Manual therapy, such as lymphatic drainage and joint mobilizations may also be used to reduce edema and improve ROM and weight bearing ability (Martin et al., 2013). A therapeutic exercise program involving ROM, strength, and balance training should be started as soon as possible following the acute phase of injury (Kaminski et al., 2013). In grade three sprains, complete immobilization or casting of the ankle and lower leg may be necessary (Martin et al., 2013). Immobilization should last for a minimum of 10 days, after which therapeutic exercises should be implemented (Kaminski et al., 2013). Limited evidence exists both for (Wilson, 1972) and against (Man, Morrissey, & Cywinski, 2007) the use of electrotherapeutic modalities.

Because ankle sprains and injuries are relatively common, many people do not see a healthcare provider following what he/she believes to be an ankle sprain (Robbins & Waked, 1998). Furthermore, it has been reported that many athletes do not follow a proper rehabilitation program, returning to play when he/she is still experiencing symptoms, or with the assistance of an ankle brace or tape (Stasinopoulos, 2004). This poses a significant danger to athletes; the risk for re-injury is greatly increased up to 12 months post injury, as the injured ligaments proceed through the phases of soft tissue healing and return to their pre-injury strength (Bahr & Bahr, 1997). As such, the greatest predisposing factor to an ankle sprain is a previous ankle sprain (Beynnon, Murphy, & Alosa, 2002).

Chronic Ankle Instability

Repeated trauma to the ligamentous structures of the ankle can lead to chronic ankle instability (CAI). By definition, CAI refers to constant lateral ankle instability, which can result in many ankle sprains over time (Hertel, 2002). Chronic ankle instability can be further divided into mechanical instability and functional instability. Mechanical instability refers to instability that is the result of ligamentous laxity following an injury (Tropp, Odenrisk, & Gillquist, 1985). In comparison, functional ankle instability generally refers to the feeling of giving way in the ankle (Freeman, 1965), characterised by deficits in proprioception and neuromuscular activity (Hertel, 2000).

Mechanical CAI is often attributed to pathological laxity of the talocrural and subtalar joints (Hertel, 2002; Martin, Kaplan, Kahler, Dussault, & Randolph, 1996) and results from ligamentous injury to the ATFL and CFL (Rasmussen & Tovborg-Jensen, 1982). As the ATFL and CFL function to provide mechanical stability to the talocrural and subtalar joints (van den Bekerom et al., 2008), damage to these structures, especially complete ruptures, can lead to instability. In turn, this can increase the risk of injury at the ankle (Boardman & Liu, 1997; Hertel, 2002), alter kinematics of proximal structures during jumping (Gribble & Robinson, 2009) and hopping (Gribble, Hertel, & Deneger, 2007), and alter lower extremity muscle activation (Feger, Donovan, Hart, & Hertel, 2014; Feger, Donovan, Hart, and Hertel, 2015).

Resulting from ligamentous laxity, arthrokinematic and osteokinematic impairments could lead to mechanical CAI. Mulligan (1995) suggested that following an acute inversion ankle sprain, the lateral malleolus and fibula may sublux anteriorly and inferiorly. This hypothesis is supported by Hubbard and Hertel (2008) and Mavi, Yildirim, Grunes, Pestamalci, and Gumusburun (2002). Since the ATFL originates on the lateral malleolus and inserts onto the talus, the resultant anterior and/or inferior displacement of the lateral malleolus creates laxity in the ATFL by bringing the origin closer to the insertion, allowing for greater inversion of the ankle. This malpositioning can lead to frequent episodes of instability and repeated ankle injury. As such, this may place other lower quadrant structures in vulnerable positions (Mulligan, 1995).

In addition to malpositioning of the fibula and lateral malleolus, restricted dorsiflexion has been demonstrated in persons diagnosed with CAI (Drewes, McKeon, Kerrigan, & Hertel, 2009; Hoch, Staton, Mckeon, Mattacola, & McKeon, 2012) and persons with previous lower body injuries (Wiesler, Hunter, Martin, Curl, & Hoen, 1996). Green, Refshauge, Crosbie, and Adams (2001) inferred that changes in the movement of the talus on the tibia could be the cause of this observed restriction in dorsiflexion. The implication of reduced dorsiflexion is that the talus is unable to tightly fit within the ankle mortise, allowing the talus to invert and the tibia and mid foot to internally rotate with reduced mechanical resistance (Hertel, 2002; Thonnard, Bragard, Willems, & Plaghki, 1996). By not providing adequate mechanical resistance to inversion and rotation, the mechanism for a lateral ankle sprain may be enhanced.

While no universal definition of functional CAI exists, it is generally referred to as instability during dynamic tasks resulting from neuromuscular deficits (Konradsen, 2002). During gait, altered lower extremity motor recruitment patterns have been identified in ankles with CAI. Santilli et al. (2005) examined muscle activation of the PL muscle in athletes with

CAI, comparing each participant's unstable ankle to his/her healthy ankle. A significant decrease in PL activity during the stance phase of gait was recorded for the unstable ankle. Feger et al. (2015) also examined PL, as well as tibialis anterior (TA), lateral gastrocnemius (LG), biceps femoris (BF), rectus femoris (RF) and gluteus medius (GM) muscle activation during gait in ankles with CAI and the corresponding lower limb. Unlike Santilli et al., Feger et al. compared persons with CAI to a matched control group with no ankle instability. In contrast to Santilli et al., PL was significantly more active during the stance phase in unstable ankles, compared to healthy ankles. Additionally, activation of the PL and RF occurred significantly earlier during the stance phase. Although not significant, TA, LG, BF, and GM were also activated earlier in the CAI group, relative to the healthy control group. While these studies present contradictory findings, both observed significant differences in muscular activation patterns in ankles with CAI, suggesting that unstable ankles can influence motor programing and muscular activation of the lower extremity during gait.

Alterations in muscular activation patterns of the lower extremity in persons with CAI have also been observed during balance, lunging, and hopping tasks. Feger et al., (2014) had participants with and without CAI perform a forward lunge exercise, lateral hopping exercise, single leg-eyes closed balance test, and the Star Excursion Balance Test (SEBT) while monitoring activation of the TA, PL, LG, BF, RF, and GM. During the lunge exercise, there was a decrease in total reflexive muscle activity of the corresponding limb after ground contact in persons with CAI. A similar decrease was observed during the SEBT and single limb balance test. During the forward lunge and lateral hopping exercises, a total decrease in preparatory muscular activation prior to ground contact was also observed in the CAI group, further indicating that CAI can affect muscular activation and motor programming.

In addition to reduced muscular activation of the unstable ankle and corresponding lower limb, alteration in muscular activation of the unaffected limb has been reported in persons with a history of ankle sprains. Bullock-Saxton, Janda, and Bullock (1994) examined muscular activation of the hamstrings, gluteus maximus, and erector spinae muscles during prone hip extension in persons with and without a history of grade one and grade two ankle sprains. Compared to the healthy control group, the onset of gluteus maximus activation was significantly slower on both the injured and uninjured sides. The authors attributed this delay in gluteus maximus muscle activation to altered mechanoreceptor input, suggesting that altered afferent feedback from the ankle could influence motor planning and muscle recruitment further up the kinetic chain.

Mechanoreceptors are sensory structures that detect changes in speed and force of a movement, providing afferent feedback to the central nervous system (CNS). In muscles, these include muscle spindles and Golgi tendon organs. Muscle spindles primarily serve to detect changes in muscle length during movement (Powers & Howley, 2012). Therefore, muscle spindles may play a key role in generating stretch reflex responses; more specifically, reflexes designed to contract lower extremity muscles to maintain foot posture when on an uneven surface (Freeman, Dean, & Hanham, 1965). Damage to the mechanoreceptors in the ankle from an ankle sprain have been hypothesized to cause proprioceptive deficits in the ankle (Konradsen & Ravn, 1990) in addition to increasing postural sway (Konradsen & Ravn, 1991). Furthermore, the loss of eversion strength due to a reduction in motor unit recruitment has been observed when the peroneal muscles are torn following an inversion ankle sprain (Kleinrensink et al., 1994). As such, current recommendations for CAI involve addressing functional deficits (i.e., strength, proprioception, balance), while promoting typical movement patterns (i.e., walking, running, stepping; Kamiski et al., 2013). When used in conjunction with a functional rehabilitation program, ankle braces are thought to increase afferent feedback from sensory structures in the foot (Feuerbach, Grabiner, Koh, & Weiker, 1994), and are often worn by persons with CAI to provide additional mechanical stability to the ankle during movement (Barlow, Donovan, Hart, & Hertel, 2015).

Ankle Braces and Taping

Taping of the ankle joint is a common practice in sport as a treatment and preventative measure for ankle sprains (Paris, 1992) and represents the most popular method of ankle injury prevention (Stasinopoulos, 2004). Ankle taping, however, is expensive and requires qualified personnel to apply (Bot & van Mechelen, 1999; Burks, Bean, Marcus, & Barker, 1991). Despite this, it is not uncommon for teams in jumping sports to have mandatory bracing or taping policies (Pedowitz, Reddy, Parekh, Huffman, & Sennet, 2008). It is estimated that, at Lakehead University alone, up to $300 CAD per basketball player can be spent annually on athletic tape; in addition to hundreds of athletic trainer hours spent taping. Such expenses are not restricted to Lakehead University; in 1991, an estimated $16,000 USD was spent to tape the entire University of Utah football player's ankles for one season (Burks et al., 1991). Adjusted for inflation, this would equal $29,316 USD in 2018. The overall cost of ankle taping represents a significant budgetary and time constraint on both institutions and their personnel. As such, reusable ankle braces have become a popular alternative to ankle taping (Frey, Feder, & Sleight, 2010; Pienkowski, McMorrow, Shapiro, Caborn, & Stayton, 1995).

Several models of ankle braces are available commercially (Frey et al., 2010). Most ankle braces are classified as either a softshell ankle brace or a semi-rigid ankle brace. Softshell ankle braces, such as the ASO® EVO® (AE; Figure 4), generally have a lace-up design and are built of a nylon material. Additional features may, or may not, include heel-lock or horseshoe straps (Gudibanda & Wang, 2005). Semi-rigid ankle braces, such as the Active Ankle T1™ (T1; Figure 5) have a hinge that sits underneath the heel, while the medial and lateral aspect of a semi-rigid brace is composed of a rigid shell (MacKean, Bell, & Burnham, 1995). Theoretically, this allows for unrestricted movement in the sagittal plan, while restricting movement in the frontal plane (Mackean et al., 1995). As such, there are structural differences between brace styles that affect how they function. Despite their structural differences, both styles of braces are commonly used across sports (Stasinopoulos, 2004).

Figure 4. ASO® EVO® softshell ankle brace. This figure displays an example of a softshell brace.

Figure 5. Active Ankle T1™ semi-rigid ankle brace. This figure displays an example of a (semi-) rigid ankle brace.

Protective mechanisms. Since ankle sprains occur when the ligament(s) of the ankle are overstretched (Ferran & Maffulli, 2006), restricting the ROM of the ankle region is considered the main mechanism by which ankle braces prevent ankle sprains and injuries (Verhagen & Bay, 2010). Additionally, improved proprioception has also been suggested as a protective mechanism (Feuerbach, Grabiner, Koh, & Weiker, 1994). As such, the ability for ankle braces to restrict ROM and improve proprioception at the ankle has been well studied in healthy and CAI populations.

Range of motion restriction. Range of motion can be divided into two categories: Active ROM and passive ROM. Active range of motion refers to motion at a joint that is generated via muscular contraction, whereas passive ROM involves another individual moving a joint without any assistance from the individual being tested (Anderson & Parr, 2013)

In controlled, static environments, both softshell and semi-rigid ankle braces have demonstrated the ability to restrict passive ankle ROM. Gross, Bradshaw, Ventry, and Weller (1987) measured passive frontal plane motion after standard athletic taping of the ankle and when wearing the Aircast Airstirrup™ semi-rigid ankle brace. As measured with an isokinetic device, passive ROM in the frontal plane was significantly reduced by 28% in healthy individuals after the application of athletic tape. After 10 minutes of continuous running, however, the ability for athletic tape to reduce passive inversion ROM at the ankle was insignificantly reduced by 5°. In comparison, ankle bracing was able to restrict passive ROM in the frontal plane by 38% immediately after application. A similar reduction was present after 10 minutes of continuous running, suggesting that a semi-rigid ankle brace may maintain its ability to restrict passive inversion ROM at the ankle after exercise better than athletic taping.

Greene and Roland (1989) investigated the effects of the Donjoy® Ankle Ligament Protector semi-rigid ankle brace on active ankle ROM. Healthy individuals with no recent history of ankle sprains had his/her active ROM in the frontal plane assessed with and without the semi-rigid ankle brace. As measured with an isokinetic device, total active ROM in the frontal plane was reduced significantly by 30%. This mean reduction in total active ROM was also observed after a 20 minute dynamic exercise session, indicating that the ability for the semi-rigid ankle brace to restrict ROM did not decrease after exercise. More recently, Tang, Wu, Liao, and Chan (2010) measured the ability of the Aircast Airstirrup™ semi-rigid ankle brace to restrict motion during a 30° perturbation. Participants stood in weight bearing on an inversion platform while either the left or right side of the platform unexpectedly dropped to 30°, creating a dynamic supination of the foot. In healthy participants, ankle inversion displacement was significantly reduced by 6° during the perturbation, compared to when he/she was barefoot. Additionally, plantarflexion displacement significantly decreased by 3° when wearing the ankle brace. As such, in persons with no reported instability of the ankle, semi-rigid ankle reduced motion in the frontal and sagittal planes.

In persons with CAI, Eils et al. (2002) compared four softshell and six semi-rigid ankle braces and their ability to restrict passive and dynamic ankle ROM. Using an isokinetic device, semi-rigid and softshell ankles braces significantly reduced passive ankle ROM in the frontal plane compared to the no ankle brace control condition. A much larger reduction was evident in the semi-rigid ankle brace models compared to the softshell ankle brace models. Similar reductions of passive ankle supination and pronation ROM was present in both brace types. Although passive ankle ROM in the sagittal plane was reduced with both styles of braces, the semi-rigid ankle brace models reduced plantarflexion and dorsiflexion motion more than the softshell ankle braces. This is surprising, given that most semi-rigid ankle braces are designed to allow for unrestricted plantarflexion and dorsiflexion (MacKean et al. 1995). As expected, semi-rigid ankle brace models were more effective for reducing supination during a sudden ankle perturbation than their softshell counterparts. While semi-rigid ankle braces restricted ankle ROM more than softshell ankle braces, none of the softshell ankle brace models incorporated heel-lock or horseshoe straps. Since the purpose of the heel-lock and horseshoe straps is to further reduce sagittal and frontal plane motion of the ankle, a greater restriction of movement may have been observed in softshell ankle braces with heel-lock and horseshoe strap designs.

A softshell ankle brace with heel-lock straps was included in Alfuth, Klein, Koch, and Rosenbaum's (2014) investigation of dynamic and passive stabilization when wearing an ankle brace. Healthy participants had his/her ankle supination measured with an in-shoe goniometer during an unexpected 30° ankle perturbation. The maximum supination angle when not wearing the ankle brace was 41.8°. When wearing the Aircast AirGo™ softshell ankle brace, the maximum supination angle was significantly reduced to 24.6°. The maximum supination angle was reduced to 27.1° and 23.6°, respectively, when wearing the Darco Body Armor® Embrace and McDavid Ankle X™ semi-rigid braces, respectively. Passive ROM was also assessed using an isokinetic device; all braces significantly reduced passive ROM in the frontal plane compared to the no ankle brace control condition. A significant reduction of passive internal rotation was observed when wearing all three ankle braces. Additionally, passive plantarflexion was significantly reduced when wearing the softshell and semi-rigid ankle braces, with the softshell ankle brace providing the most passive restriction in the sagittal plane. When compared to Eils et al. (2002), it appears that incorporation of a heel-lock strap in the design of a softshell ankle brace can significantly improve the restrictive properties in both the sagittal and frontal planes. Although this study only examined healthy individuals and one softshell ankle brace model, similar restrictive properties in a softshell ankle brace were reported by Miller, Needle, Swanik, Gustavsen, and Kaminski (2012).

Miller et al. (2012) compared the passive ankle ROM restriction capabilities of the AE softshell ankle brace and standard athletic taping, both forms of external ankle supports that incorporated heel-lock strap designs. To compare ankles with and without CAI, both ankles of each participant were categorized by the score on the 30-point Cumberland Ankle Instability Tool (CAIT) and self reported ankle sprain history. Ankles were assigned to either the healthy control group (CAIT score greater than 28, no previous ankle sprains), previous history group (CAIT score greater than 28, previous ankle sprain), or unstable ankle group (CAIT score less than 24, previous ankle sprains). Using an isokinetic device, participant's maximum inversion, eversion, and anterior displacement were measured after the application of an external support, after 20 minutes of jumping, running, and agility exercise, and after the removal of the external ankle support. In all groups, the AE ankle brace and standard athletic taping significantly restricted inversion and eversion motion, as well as anterior displacement before exercise. This reduction was also present after exercise; however, it was slightly reduced compared to the pre-exercise values. As such, it appeared that external devices that incorporated heel lock straps to restrict sagittal plane motion were able to maintain this restriction after the completion of the exercise.

The aforementioned studies examining ankle bracing and ROM restriction lend support to the notion that ankle braces significantly reduce passive and active ROM in all planes, as well as supination movement of the ankle during a perturbation. Results from Simpson, Cravens, Higbie, Theodorou, and Delrey (1999), however, suggested that this may not be the case. During a lateral cutting task, plantarflexion was reduced when wearing the Aircast Airstirrup™, MalleoLoc® semi-rigid ankle brace, and the Swede-o®® softshell ankle brace compared to the no brace control condition. Despite this, no significant differences were observed in the dynamic supination angle between the three ankle braces and the no brace control condition. As all study participants had at least one previous ankle sprain, the authors suggested that the lack of reduction in supination angle could have been the result of participants being hesitant about performing the task in the no ankle brace control condition. Despite the results of this study, it appears that ankle braces are effective for restricting ROM at the ankle.

Improved proprioception. Another suggested mechanism by which ankle bracing decreases the risk of injury is by improving proprioception at the ankle (Olmsted, Vela, Denegar, & Hertel, 2004). Although definitions vary, proprioception generally refers to the awareness of the body or a body part, such as the ankle, in space (Johnson & Soucacos, 2010). As such, proprioception is heavily reliant on afferent feedback from receptors in muscle, tendons, and skin during movement (Lephart, Pincivero, & Rozzi, 1998). Freeman et al. (1965) suggested that, after an ankle sprain, the mechanoreceptors of the ankle are damaged. As such, this leads to improper ankle joint positioning, placing the ankle in a vulnerable position for further injury. By applying an ankle brace, it is thought that cutaneous receptors in the contacted skin area become more active, increasing afferent feedback to the central nervous system, resulting in improved joint positioning awareness (Feuerbach et al., 1994).

The effect that a semi-rigid ankle brace compared to anesthetizing the ATFL and CFL on joint repositioning capabilities was investigated by Feuerbach et al. (1994). The right (dominant) foot in healthy participants was passively placed into a total of nine different positions, incorporating all three planes of motion. After being passively placed into each position by the researcher, participants were then required to actively recreate the position. Participants completed the protocol without an ankle brace and then repeated it when wearing the Aircast Airstirrup™ semi-rigid ankle brace. Participants then repeated the protocol with the ATFL and CFL anesthetized. When wearing a semi-rigid ankle brace without having the ATFL and CFL anesthetized, absolute error in joint position replication was significantly reduced, supporting the notion that mechanoreceptors in the ankle contributed to joint positioning awareness when stimulated by an ankle brace. No difference in absolute error, however, was observed following the anesthetizing the ATFL and CFL. As such, mechanoreceptors in the lateral ligament complex may not contribute significantly to ankle proprioception. Rather, the authors suggested that stimulation of the cutaneous receptors in the skin may be responsible for improved ankle proprioception when wearing an ankle brace.

The results of Feuerbach et al.'s (1994) study are also supported by Heit, Lephart, and Rozzi (1996). The ability of healthy participants to actively reproduce ankle positions with the right ankle was examined when wearing the Swede-o® softshell ankle brace compared to the application of standard athletic taping of the ankle and when barefoot. Participants were passively moved into either 30° of plantarflexion or 30° of inversion using an isokinetic device. Participants where then required to actively recreate the position. The use of a Swede-o® ankle brace significantly improved the participant's ability to reproduce active plantarflexion angle, yet there was no significant improvement in his/her ability to reproduce the inversion angle. Again, the authors hypothesized that this improvement in joint position was the result of enhanced stimulation of the cutaneous receptors in the skin and mechanoreceptors in the ankle ligaments. This effect may be specific to plantarflexion, however, as the softshell ankle brace failed to improve inversion position reproduction.

Ankle Bracing and Injury Prevention

Irrespective of the mechanism by which ankle braces prevent ankle injuries, the current epidemiological evidence suggests that, overall, they are effective. Recent systematic reviews by Leppänen et al. (2014) and Petersen et al. (2013) concluded that in the athletic populations, wearing ankle braces reduced the overall risk of ankle injuries. As such, the NATA currently recommends that all athletes returning to play from an ankle injury wear ankle braces (Kaminski et al., 2013). While the majority of epidemiological studies support wearing ankle braces to reduce the risk of ankle injury, a few studies have not come to this conclusion. Furthermore, it appears that the ability for ankle braces to reduce the risk of ankle injury may be dependent upon whether or not the person has a prior history of ankle injury.

In a sample of 52 female volleyball players with and without a prior history of ankle sprains, Stasinopoulos (2004) investigated the effectiveness of three different interventions for reducing the incidence of ankle sprains: technique training, proprioceptive training, and wearing the Aircast Airstirrup™ semi-rigid ankle brace. The technique training focused on attacking and blocking approaches with an emphasis placed on the final approach step and ensuring that players jumped straight up when attacking the ball. Proprioceptive training had players perform exercise on a balance board for 30 minutes per day. The ankle brace group was instructed to wear the Aircast Airstirrup™ ankle brace bilaterally for all games and practices. All three interventions reduced the number of ankle sprains, with technique training being the most effective and the Aircast Airstirrup™ ankle brace being the least effective. Despite this, in players who sustained more than four ankle sprains in his/her career, the Aircast Airstirrup™ ankle brace did not reduce the incidence of ankle sprains. In comparison, technique and proprioceptive training still decreased the number of ankle injuries in persons who sustained more than four ankle sprains. As such, the results of this study suggested that neuromuscular deficits may arise from repeated ankle sprains, and that addressing these deficits may be more beneficial for preventing repeated ankle sprains than the use of ankle braces.

Frey et al. (2010) conducted a prospective comparative study examining the effect of ankle bracing on ankle injury rates in 999 high school volleyball players. A total of five softshell and semi-rigid braces were examined. Compared to players who did not wear ankle braces, players who wore ankle braces did not experience a significant reduction in ankle injuries. When comparing male and female participants, there were significantly more injuries in female players who wore softshell ankle braces, compared to males wearing the same softshell ankle brace. Additionally, there were significantly more ankle injuries in women who wore semi-rigid ankle braces, compared to women who wore softshell ankle braces. Furthermore, when controlling for participants who had a previous history of ankle injury, the Active Ankle T2™ and Aircast Airsport™ semi-rigid ankle braces only reduced the risk of ankle injury in persons without a previous history of ankle injury. Based on this study, it appears that sex and brace style may influence the effectiveness of ankle braces for reducing ankle injuries. The results of this study also lend support to Stasinopoulos (2004), in that the ability of ankle braces to reduce ankle injuries may be decreased in persons who sustained previous ankle sprains or injuries. There is research, however, to suggest that ankle braces are only effective in persons with a previous history of ankle sprains.

Surve, Schwellnus, Noakes, and Lombard (1994) conducted a randomized prospective clinical trial examining the relationship between Aircast Airstirrup™ semi-rigid ankle braces and ankle injuries in competitive soccer players. In persons with no history of ankle sprains (n = 246), compared to the no prior history and no brace control group, the Aircast Airstirrup™ ankle braces did not reduce the number of ankle sprains during one season. Conversely, in persons with a prior history of ankle sprains (n = 258), wearing the Aircast Airstirrup™ ankle braces significantly reduced the incidence of ankle sprains compared to the prior history and no brace control group. In addition to recording the incidence of ankle sprains, Surve et al. also documented the severity of all ankle injuries that occurred during the study. There was no difference in the severity of ankle sprains between the no history ankle brace group and the control group. Persons with a prior history of ankle sprains, however, who wore semi-rigid ankle braces experienced fewer severe (grade two and three) ankle sprains than the control group.

Surve et al. (1994) noted that semi-rigid ankle braces were more effective for decreasing injury severity in persons who had a prior history of ankle sprain than persons who did not. The same cannot be said for softshell ankle braces. McGuine, Hetzel, Wilson, and Brooks (2012) conducted a randomized controlled trial of 2,081 high school basketball players, comparing the incidence and severity of lower extremity injuries in persons who wore softshell ankle braces to persons who did not. Over the course of a full season, persons who wore AE ankle braces experienced significantly less ankle injuries than those who did not wear ankle braces. Despite reducing the number of ankle injuries, however, softshell ankle braces did not reduce the severity of ankle injuries. Unlike Surve et al., these differences were still present when controlling for a prior history of ankle injuries and sex. Additionally, incidence and severity of lower extremity injuries were similar between the ankle brace and no ankle brace control conditions.

McGuine et al. (2012) did not report any differences in lower extremity injuries between the softshell ankle brace and no ankle brace control conditions, although this was not the result of a previous study. McGuine, Brooks, and Hetzel (2011) conducted a similar study of high school basketball players (n = 1460) and noted that AE ankle braces significantly reduced ankle injuries compared to the no ankle brace control condition. The severity of ankle injuries was also similar between ankle brace and no ankle brace conditions. Again, these results did not change significantly when controlling for prior ankle injury and sex. Unlike McGuine et al. (2012), however, there was a non-significant 85% increase in lower extremity injuries that were not ankle related when wearing a softshell ankle brace. The majority of these injuries were acute musculoskeletal strains of the lower leg, upper leg, and hip. As such, McGuine et al. (2011) suggested that further research be done on the effects of ankle braces on lower extremity and lumbar spine biomechanics.

Although McGuine et al. (2011) found that ankle braces were effective for reducing ankle injuries, it is not the first study to suggest that wearing prophylactic equipment may increase the risk of injury. Yang et al. (2005) conducted a 3 year prospective cohort study examining prophylactic equipment use in high school athletes and its relation with lower extremity injury rates and severity. A total of 19,728 athletes from 12 different sports were surveyed regarding his/her use of prophylactic equipment such as knee braces, knee pads, and ankle braces. Athletes were also asked about their previous history of injury. Injury and exposure reports were filled out weekly by designated data collectors at the athlete's institution. Overall, lower extremity injury rates were reduced when wearing prophylactic equipment, especially when wearing knee pads. Despite this, when controlling for the type of equipment, knee braces and ankle braces were associated with a greater risk of any lower extremity injury. The authors speculated that the reduction in injuries when wearing knee pads, compared to knee and ankle braces were the result of design; knee pads provide protection without the use of rigid materials or significant ROM restriction, whereas braces can often incorporate rigid materials and restrict ROM. Furthermore, braces may slip out of position, unfavourably altering their function. While it is important to acknowledge the many limitations of this study (i.e., self reported, consistency of prophylactic equipment use, intensity of practice), given the results of McGuine et al. (2011), more research investigating the effects of prophylactic equipment use on the biomechanics of the lower extremity and injury is warranted.

Biomechanics of Ankle Bracing

Biomechanically, much is known about the regional effects of wearing ankle braces. As previously mentioned, restriction of ankle ROM is considered the primary protective mechanism of an ankle brace (Verhagen & Bay, 2010). The effects of restricting normal ankle ROM on kinetics, kinematics, and electromyography (EMG) of the lower extremity, however, are not well understood. During a jump landing, many biomechanical factors have been associated with reduced ground reaction forces (GRFs). These factors include: larger knee flexion angles (Dufek & Bates, 1990; Wernli, Ng, Phan, Davey, & Grisbrook, 2016), lower extremity extensor and flexor, eccentric and concentric contraction (Devita & Skelly, 1992), and increased dorsiflexion and plantarflexion ROM (Gross & Nelson, 1988). As such, both kinematic and EMG variables may play a role in reducing the GRFs. How these variables are affected by an ankle brace remains unclear. Decreased quadriceps muscle activity and increased soleus muscle activity was observed when plantarflexion was artificially reduced during the eccentric portion of a squat (Macrum, Bell, Boling, Lewek, & Padua, 2012). Given that ankle braces have been shown to restrict dorsiflexion and plantarflexion, similar changes in lower extremity muscular activation, as well as kinematic changes may be present during other athletic tasks. As such, the body's ability to naturally attenuate GRFs during a jump landing or cutting task may be affected. Furthermore, this may also lead to reductions in performance during athletic performance measures, such as when assessing vertical jump height. Despite this, few studies have explored the kinetics and kinematics when wearing ankle braces during jumping, cutting, and agility tasks or maneuvers. Furthermore, no studies have examined the effects of ankle braces on proximal lower extremity muscular activation during jumping, cutting, agility tasks, or maneuvers.

Kinetics. In biomechanics, kinetics refers to the study of forces such as shear, torque, and GRFs (Hall, 2012). Much of the literature examining the kinetics of wearing ankle braces has investigated the effects on GRFs.

Ground reaction forces. Ground reaction forces are based on Newton's Third Law of Motion; for every action, there is an equal and opposite reaction (Hall, 2012). Applied to the human body, GRFs represent the force exerted by the ground on bodily contact, measured in vertical (F_z), anteroposterior (F_y), and mediolateral planes (F_x; Nilsson & Thorstensson, 1989). It is theorized that if external loads and forces, such as GRFs, exceed that which can be dissipated by the musculoskeletal system, injury to the lower extremity may occur (Dufek & Bates, 1990). This makes the magnitude, as well as the time over which the force is applied (impulse) important from an injury perspective. When landing from a 30 cm jump, GRFs have been shown to be an average of 4.6 bodyweights greater than standing (McNair & Prapayessis, 1999). During normal running gait, GRFs during foot impact have been shown to be 1.6 to2.3 times that of bodyweight, while GRFs during the propulsive phase averaged 2.5 to 2.8 times that of bodyweight (Munro, 1987). Given the magnitude of GRF generated during these tasks, effectively attenuating these forces may play a key role in preventing lower extremity injury.

Jumping. Recently, Castro, Goethel, Gáspari, Crozara, and Gonçalves (2017) investigated GRFs during a basketball rebound jump in 11 adolescent basketball players when wearing bilateral Horse Jump® softshell ankle braces. After performing a vertical jump to a standardized height for each participant, a basketball was hung overhead, to approximately 95% of the vertical jump height. In a countermovement fashion using both hands, participants were instructed to jump up and touch the basketball. To measure GRFs, participants performed each jump on a force platform. Each participant performed the task with and without bilateral Horse Jump® ankle braces. During takeoff, Horse® jump ankle braces did not affect peak vertical GRF or vertical impulse, nor were lateral and medial GRF peaks affected. A significant reduction, however, was observed upon landing in lateral and medial GRF peaks, but not vertical peak GRF or impulse.

Riemann, Schmitz, Gale, and McCaw (2002) also investigated GRFs during a jump when wearing ankle braces. Unlike Castro et al. (2017), Riemann et al. utilized a single leg drop landing, in addition to an exercise bout to gauge the ability for ankle braces to maintain any potential effects on GRFs after usage. Fourteen physically active college students performed a stiff and soft single leg drop landing under three bilateral conditions on 3 separate days: no ankle support control, Aircast Airstirrup™ semi-rigid ankle brace, and standard athletic taping (modified Gibney technique) of the ankle. Participant's then ran on a treadmill for 20 minutes at 65-70% of his/her heart rate reserve, before performing the landings once again. When wearing the Aircast Airstirrup™ ankle braces, the time to maximum vertical GRF was significantly less than the no ankle support control condition, before and after exercise and during both types of landings. As such, while the magnitude of vertical GRF was not affected, the rate at which this peak value was reached increased. The authors suggested that this may be due to the reduction of ankle ROM provided by the application of athletic taping to the ankle, as well as the use of the semi-rigid ankle braces. Only GRF data were collected in this study, making any mechanisms behind this speculative at best. Therefore, the implications of reaching maximum vertical GRF magnitude more quickly are unknown.

Cutting tasks. Cloak, Galloway, and Wyon (2010) compared mediolateral GRFs during a sidestep cutting task with and without bilateral Aircast Airstirrup™ semi-rigid ankle braces. Ten healthy, male basketball players approached a force platform at speeds between 4.5 and 5.5 metres per second. When the dominant foot made contact with the force platform, participants cut 45° from the original plane. When wearing the Aircast Airstirrup™ ankle braces, peak mediolateral GRF was significantly reduced, compared to the no ankle brace condition. No other GRF variables were reported. de Comargo Neves Sacco et al. (2006), however, performed a similar study (n = 8) using a basketball cutting maneuver, separating the time in contact with the force plate into two phases: foot contact and propulsive. While no significant differences were found between the Aircast Airstirrup™ ankle braces, athletic taping, and no ankle brace conditions at the point of foot contact, there was a significant reduction in peak vertical GRF during the propulsion phase, as well as the time to reach peak vertical GRF. Impulse measures were not significantly different between conditions. This is interesting to note, given that GRF is related to the ability to produce power in the lower extremity (Cordova & Armstrong, 1996). Although time to complete the maneuver was not recorded, decreased GRF at the propulsive phase of cutting may be indicative of reduced power, thus impacting on speed and performance. Without data concerning time to peak GRF, however, this is speculative and only represents one possibility. Therefore, the mechanisms behind any such change in GRF during the propulsive phase or performance remains unclear without additional kinetic and/or kinematic data.

Kinematics. In biomechanics, kinematics describes motion without reference to the forces causing motion. Linear and angular kinematic variables include distance, displacement, acceleration, and velocity. (Hall, 2012). The majority of research examining lower extremity kinematics when wearing ankle braces has focused on position and movement of the hip and knee. An even larger focus has been placed on the kinematics of the knee when wearing ankle braces, as large internal and external rotation ROM at the knee joint (Noyes, Mooar, Matthews, & Butler, 1983), increased knee extension (Podraza & White, 2010), and knee abduction angle (Hewett, Torg, & Boden, 2009) has been associated with anterior cruciate ligament injury (ACL). As such, if ankle braces were to increase these variables, mechanisms of ACL injury may be promoted.

Kinetics and kinematics of ankle bracing.

*Jump landings***.** Landing from a jump is a physical task required in many sports and is one of the predominant mechanisms of noncontact knee (Boden et al., 2000) and ankle injuries (Caulfield & Garret, 2004; McKay et al., 2001). In kinetic and kinematic analyses of the knee and hip when wearing ankle braces, results have been mixed. Hodgson, Tis, Cobb, and Higbie (2005) examined the effect of semi-rigid ankle braces on lower extremity kinematics and kinetics during a simulated, two foot jump landing. Twelve healthy female collegiate volleyball players, all of which had experience wearing the Active Ankle T2™ semi-rigid ankle brace, suspended themselves 0.61 metres above a force platform. This was accomplished by having the participant hang from an adjustable bar, from which the participant would let go and land with the left foot in contact with the force platform. Participants completed two drop landings with and without bilateral Active Ankle T2™ ankle braces. No significant differences were found in knee and hip kinematics when completing the task with or without Active Ankle T2™ ankle braces. Peak vertical GRF and loading rate during toe contact, however, was significantly greater when wearing the Active Ankle T2™ ankle braces. This was accompanied by a significant decrease in ankle angle change at touchdown. As such, it would appear that the Active Ankle T2™ ankle braces restrict ankle ROM during landing, potentially increasing vertical GRF by reducing total displacement of the ankle.

Similar results were noted by Simpson et al. (2013) using softshell ankle braces. Sixteen healthy, athletic females suspended themselves 0.43 metres above the ground by hanging on to an adjustable bar. Participants then let go of the bar and landed with the dominant foot on the force platform. Participants completed the hanging drop with and without bilateral ASO® softshell ankle braces. At landing, plantarflexion angle was significantly reduced by 8° when wearing the ASO® ankle braces. Participants also generated a significantly larger peak vertical GRF and mediolateral GRF value when wearing the ASO® ankle braces, compared to landing without ankle braces. Furthermore, the point at which maximum vertical GRF was reached was 6.4 milliseconds earlier when wearing an ankle brace. As with Hodgson et al. (2005), the ankle was in significantly less plantarflexion at touchdown compared to the no brace condition. As such, this may have resulted in larger peak vertical GRF due to the restriction of ankle ROM and displacement. Participants experienced significantly less internal and external knee rotation at landing, in addition to a significant 3° increase in knee flexion in the ASO® ankle brace condition, which hypothetically should help reduce GRFs. As semi-rigid ankle braces did not affect knee or hip kinetics (Hodgson et al., 2005), this would suggest a compensatory strategy at the knee involving increased knee flexion to make up for reduced plantarflexion and dorsiflexion provided by a softshell ankle brace, perhaps in an attempt to reduce GRFs. Therefore, it appears that softshell and semi-rigid ankle braces may increase vertical GRF, but affect proximal biomechanics in different manners.

Support for an adaptive compensatory strategy at the knee to attenuate GRFs can be found in DiStefano, Padua, Brown, and Guskiewicz's (2008) investigation of the kinetic and kinematic effects of wearing AE ankle braces during a 30 cm high broad jump landing. Thirty seven healthy, recreational basketball and volleyball players performed broad jump landings under two different conditions: with and without bilateral AE ankle braces. When horizontal distance to the force platform was standardized to half of the participant's height, ankle ROM was significantly decreased at ground contact, in line with previous research (Simpson et al., 2013; Hodgson et al., 2005). Similar to Simpson et al. (2013), knee flexion angle increased by 3° at ground contact when wearing the AE ankle braces, compared to the no ankle brace control condition. No change, however, was observed in GRF variables when wearing the AE softshell ankle braces, compared to no ankle braces, suggesting that the increase in knee flexion experienced when wearing AE ankle braces may effectively attenuate GRFs. Given these results, kinematic changes that occur when wearing softshell and semi-rigid ankle braces may be interrelated with kinetic variables. Softshell and semi-rigid ankle braces, however, may produce very different kinetic and kinematic changes during a two-foot jump landing and, therefore, requires additional investigation.

Although optimal landing technique may involve two feet, this is not always possible in a dynamic, uncontrolled game environment. Rather, athletes may be required to land on one leg, depending on the situation. The majority of ACL injuries that involve landing from a jump occur when the athlete lands on one leg in end range knee extension, creating a significant valgus force and externally rotated position of the knee (Olsen, Myklebust, Engebretsen, & Bahr, 2004). Cordova, Takahashi, Kress, Brucker, and Finch (2010) investigated the effects of the McDavid Ultra™ semi-rigid ankle brace and standard athletic taping of the ankle on lower extremity kinetics and kinematics during a 30.5 cm single leg drop landing. Thirteen healthy males completed the drop landing when wearing the McDavid Ultra™ ankle brace, standard athletic taping (basketweave technique), and with no ankle support on the dominant leg. The McDavid Ultra™ ankle brace and no ankle support conditions produced similar peak vertical GRF values at the first point of impact, however, both the McDavid Ultra™ ankle brace and taping conditions reached peak GRF significantly faster than the no ankle support condition. This effect was also present at the second point of impact. Furthermore, the McDavid Ultra™ ankle brace and taping significantly reduced ankle ROM at landing, while the semi-rigid ankle brace also significantly reduced knee displacement by 2.5°. Along with Simpson et al.'s (2013) findings, this would suggest altered knee kinematics in response to the reduction in plantarflexion and dorsiflexion; however, given the role of knee flexion in the attenuation of GRF (Devita & Skelly, 1992) and knee extension in ACL injuries, this effect may be maladaptive. Similar effects are also present in kinetic and kinematic studies of cutting tasks when wearing semi-rigid ankle braces.

Cutting tasks. The need to perform quick changes in direction is an important component of many sports (Bompa & Haff, 2009) and is a common ankle sprain mechanism (McKay et al., 2001). As such, the effect of ankle braces on kinetics and kinematics during cutting and agility like tasks has also been investigated. As with studies examining jump landings when wearing ankle braces, results have been inconsistent.

Using a 90° cutting task, Klem et al. (2016) explored the kinetic and kinematic effects of wearing softshell and semi-rigid ankle braces during a quick change in directions. To perform the cutting task, competitive female basketball players (n = 20) approached a force platform at full speed from a 45° angle. Participant's planted her left foot on the force platform before immediately cutting 90° away from the force platform. All participants performed the cutting task under three bracing conditions: no ankle braces, wearing bilateral AE ankle braces, and bilateral Active Ankle T2™ semi-rigid ankle braces. The Active Ankle T2™ ankle braces significantly restricted maximum inversion angle but not dorsiflexion, while the AE ankle braces did not affect either measure. A significant increase was observed in anterior shear and compressive forces at the knee during the deceleration phase of cutting when wearing the AE ankle braces, compared to when not wearing an ankle brace under the control condition. Conversely, anterior shear forces at the knee were significantly reduced when wearing the Active Ankle T2™ ankle braces. A significant increase in knee internal rotation was also present when wearing both the AE and Active Ankle T2™ ankle braces. Again, this study suggested that softshell and semi-rigid ankle braces may produce significant changes in lower extremity kinetics and kinematics, although semi-rigid angle braces may affect sagittal plane kinematics less than their softshell counterparts.

With respect to shear forces at the knee, West, Ng, and Campbell (2013) reported similar findings in 15 competitive volleyball players. For the cutting task component of their kinetic and kinematic evaluation of semi-rigid ankle braces, participants ran towards a force platform, before cutting 90° away from the platform. The cutting task was performed with and without bilateral

Active Ankle T2™ semi-rigid ankle braces. Not unlike Klem et al. (2016), a decrease in shear forces at the knee was observed when wearing the Active Ankle T2™ ankle brace, albeit in the lateral direction. No differences, however, were observed in lower extremity kinematics, aligning with the results of Greene, Stuelcken, Smith, and Vanwanseele (2014).

In their studies, Klem et al. (2016) and West et al. (2013) employed a 90° cutting task. Greene et al. (2014), however, used a 45° sidestep cutting task to evaluate lower extremity kinetics and kinematics when wearing a generic softshell ankle brace on the right ankle. Healthy netballers (n = 10) ran 5 metres towards a force platform; contacting the force platform with the right foot and sidestepping to the left at an angle of 45° to the approach plane. No kinematic or kinetic differences were observed between conditions, suggesting that a softshell ankle brace does not affect kinematics or kinetics during a sidestep cutting tasks.

Running. Although running for extended periods of time is a requirement of many sports, there is limited research examining the effects of ankle braces on running in the sagittal plane. Recently, however, Tamura et al. (2017) examined the effects of the ankle bracing and athletic taping of the ankle on lower extremity kinematics and total energy expenditure during treadmill running. Thirteen physically active individuals completed a continuous 30 minute treadmill run at his/her own pace under four bilateral conditions: ASO® softshell ankle brace, Active Ankle T2™ semi-rigid ankle brace, ankle tape (basketweave), and no brace/tape. Kinematic, as well as metabolic data were collected at 5 minute intervals during the 30 minutes of continuous running. All ankle support conditions significantly decreased total inversion and eversion movement, although the Active Ankle T2™ ankle brace was the only condition to significantly restrict inversion. Furthermore, the ASO® ankle brace and ankle tape significantly decreased maximum plantarflexion and velocity at toe off. Proximal to the ankle, knee kinematics were unaffected in the braced conditions, although all conditions resulted in significantly more hip abduction during initial contact than the no brace/tape condition. Finally, the Active Ankle T2™ ankle brace and tape resulted in significantly higher energy expenditure than the ASO® ankle brace or no brace/tape condition. Given that the Active Ankle T2™ ankle brace did not significantly affect plantarflexion or dorsiflexion, while the ASO® softshell ankle brace did, the increased energy expenditure seen in the Active Ankle T2™ and not the ASO® condition is surprising. As the Active Ankle T2™ ankle brace did affect maximum ankle inversion, this could indicate that pronation and supination of the foot is being affected during the gait cycle, resulting in increased hip abduction. As this increase in hip abduction was observed in in all conditions, however, additional factors such as muscle activation may influence lower extremity kinematics and GRFs when wearing ankle braces.

Electromyography. Electromyography is the measurement of neuromuscular activity of the human body and can be used to measure the quantity and timing of muscle contraction (Criswell, 2010). Electromyography of regional ankle muscles, such as the PL muscle, while wearing an ankle brace during a perturbation task has been well studied. The EMG activity and effect on the proximal lower extremity muscles when wearing ankle braces during functional tasks, however, has not been explored.

Relatively little research has been done on normal EMG characteristics of a vertical jump as well. When examining the effects of internal and external focus on vertical jump height, Wulf, Dufek, Lozano, and Pettigrew (2010) noted that root mean square error values of anterior lower extremity muscles (RF, vastus lateralis, TA) were greater than that for the posterior muscles (BF, LG) during takeoff. When performing a vertical jump, the onset of muscle activation during takeoff has been shown to occur in a proximal to distal manner relative to the joint that the muscle acts on. The gluteal and quadriceps muscle groups generate the majority of power during a vertical jump, which is then transferred to the ground via the biarticular gastrocnemius and smaller plantar-flexors. These muscles, however, will also generate power to a degree and have been shown to be responsible for as much as 25% of vertical jump height (Pandy & Zajac, 1991). As such, the activation of proximal and distal musculature is required for optimal jump performance (Pandy & Zajac, 1991). Upon landing, the quadriceps muscles activation is 1.5-5 times greater than that for the hamstring muscles, based on a percentage of maximal voluntary contraction (MVC). Furthermore, males will contract the hamstring muscles at a significantly higher percentage of MVC than females at the point of landing (Urabe et al., 2005).

At foot strike, cutting tasks are characterised by quadriceps muscle activation beyond MVC levels, submaximal hamstring activation, and a knee flexion angle of approximately 22° (Colby et al., 2000; Houck, 2003). High quadriceps muscle activation, however, can result in an increased anterior tibial displacement and shear forces at the knee, potentially increasing the risk for ACL injury (Colby et al., 2000). Therefore, the simultaneous co-activation of the antagonistic hamstring muscles is necessary to help support the ligaments of the knee and maintain stability of the joint (Draganich, Jaeger, & Kralj, 1989). As with jumping, this makes the timing, as well as the magnitude of muscle activation an important component of performance and injury prevention, especially in the female population, who exhibit lower hamstring muscle activation (Malinzak, Colby, Kirkendall, Yu, & Garret, 2001).

Landing. Considering the importance of proximal to distal force transfer, as well as magnitude and timing of muscular contraction for performance and injury prevention, research examining these variables when wearing ankle braces is scarce. Hopper, McNair, and Elliot

(1999) conducted the only study to investigate the effects of the Swede-o® softshell ankle brace on vertical GRFs and EMG activity of the TA, gastrocnemius, and PL muscles during a single-leg, height standardized landing task. Compared to landings with no braces, the Swede-o® ankle brace produced significantly less EMG activity in the gastrocnemius and PL muscle groups, while peak vertical GRF was unaffected. Although kinematic data of the lower extremity was not collected, this study provides evidence that EMG activity of the lower extremity may be affected by wearing ankle braces during a landing task, although peak vertical GRFs were not negatively affected. Given the changes in knee kinematics observed by DiStefano et al. (2008), Klem et al. (2016), and Simpson et al. (2013), and the biarticulated nature of the gastrocnemius muscle, more proximal musculature could be attenuating GRFs and requires further investigation.

Cutting. Focusing on activation of the peroneal muscles, Gribble, Radel, and Armstrong (2004) examined the effects of wearing the ASO® softshell ankle brace during a side shuffle task involving a change in direction and dynamic supination of the foot. Fifteen healthy female athletes had the EMG activity of the right PL muscle evaluated immediately after application of the brace and during a side shuffle task when wearing the ASO® ankle brace. Electromyographic activity was also collected when not wearing an ankle brace for the left PL muscle, representing the control condition. Participants were then instructed to wear the ankle brace for an additional 2 weeks, after which he/she was evaluated again. After 2 weeks of wearing the ankles brace, participants were instructed to not wear the ankle braces for 2 weeks, after which they were once again evaluated. No significant effect of time or ankle brace was noted; however, it is important to note that the authors examined EMG of the PL during the entire shuffle task, rather than when changing direction. Ankle sprains can occur when the foot is in a fixed or planted position (Ferran & Maffulli, 2006), as it is when changing direction. Thus, the effects that ankle braces have on EMG in this planted position may be more relevant with respect to injury prevention mechanisms than EMG throughout the entire task.

Cordova, Armstrong, Rankin, James, and Yeasting (1998) also examined the effects of wearing ankle braces during a lateral shuffling task, specifically EMG of the PL, TA, and LG muscles, as well as the associated GRFs. In comparison to Gribble et al. (2004), Cordova et al. (1998) examined EMG at the point of the change in direction. Twenty four healthy, active males performed a lateral shuffling task under three conditions: no ankle braces, bilateral Aircast Airstirrup™ semi-rigid ankle braces, and bilateral T1 ankle braces. For the side shuffling task, participants side shuffled at 80-90% of maximum speed towards a force platform; making contact with the force platform, participants immediately reversed his/her direction, creating a dynamic supination of the foot and ankle, and side shuffled back to the start. Compared to the control condition, wearing the Aircast Airstirrup™ nor T1 ankle braces affected GRFs at the point of contact with the force platform. There was, however, a reduction in PL EMG activity during the moment of peak impact force when wearing either semi-rigid ankle brace. The authors interpreted this reduction in PL EMG activity in two ways: 1) the reduction in PL EMG was beneficial and suggested that the ankle brace effectively reduced the stress that would normally be placed on the PL muscle, or 2) the reduction in PL EMG activity was detrimental; as the PL muscle functioned to resist inversion of the foot (Tortora & Nielson, 2010). A reduction in the PL EMG activity may then hinder a potential natural protective mechanism to ankle supination.

Perturbations. The reduction in PL EMG activity observed by Cordova et al. (1998) supports a commonly held hypothesis that wearing an ankle brace or braces may be detrimental to the natural muscular function of the ankle (Cordova & Ingersoll, 2003; Feuerbach & Grabiner, 1993). Since the PL muscle functions to evert and plantarflex the ankle (Tortora & Nielson, 2010), the PL muscle may protect the ankle during inversion movements by acting as an antagonistic muscle to ankle inversion (Konradsen, Voight, & Hojsgaard, 1997). It has been theorized that use of an ankle brace or braces, especially long term use, can reduce strength and timing at which the PL muscle contracts, as the ankle adapts to being supported by an ankle brace (Cordova & Ingersoll, 2003; Feuerbach & Grabiner, 1993). This theory has been explored using perturbations to simulate an ankle sprain.

As with other areas of ankle bracing research, the results from studies examining the EMG activity of the PL muscle during a sudden perturbation are mixed. Cordova et al. (2000) investigated both the long term and short term effects of semi-rigid and softshell ankle braces on PL muscle latency (timing) during a 35° perturbation. When wearing the T1 ankle brace or the McDavid 199™ softshell ankle brace on the dominant extremity, no differences in the timing of the PL contraction were observed in healthy individuals. Similar results were present after wearing each ankle brace for eight hours per day for eight weeks. In a comparable study, Cordova and Ingersoll (2003) found that the PL EMG activity significantly increased during a perturbation task immediately after the application of the McDavid 199™ ankle brace, and after 8 weeks of wearing the T1 ankle brace. Conversely, Shima, Maeda, and Hirohashi (2005) found that the activation of the PL muscle was delayed in both healthy and unstable ankles when wearing a generic softshell ankle brace during a 25° perturbation task. Given the discrepancies seen in PL EMG activity in studies examining lateral shuffling and perturbation tasks, more research is needed on the effects of ankle braces on the PL muscle during dynamic tasks, especially athletic performance measures.

Performance. To date, there has only been one published study that has investigated the performance, kinematic, and EMG effects of wearing ankle braces. Smith et al. (2016)

investigated the effects of a McDavid 195T™ softshell ankle brace on Vertec™ measured vertical jump height, lower extremity kinematics, and EMG activity of the soleus and gastrocnemius muscles. When wearing the McDavid 195T™ ankle brace bilaterally, varsity athletes (n = 20) experienced a significant reduction in vertical jump height of .92 in (2.33 cm), from the no ankle brace control condition. A significant reduction in ankle ROM, hip flexion, and soleus muscle activity was also observed when wearing the McDavid 195T ankle braces. The authors suggested that the reduction in soleus muscle activity was the result of decreased ankle ROM, along with a reduction in hip flexion, this decrease contributed to the observed reduction in vertical jump height. As this is the only study to investigate lower extremity EMG activity during a vertical jump, more research is needed to determine the extent to which altered EMG activity variables contributed to a decrease in vertical jump height. Many kinetic and kinematic variables, however, have been linked to vertical jump height performance. These include segmental angular displacement of the upper leg and trunk (Hsieh & Cheng, 2016), production of mechanical power (Aragón-Vargas & Gross, 1997), ankle angle at take-off (Aragón-Vargas & Gross, 1997), and minimum angle of the hip, knee, and ankle joint (Hsieh & Cheng, 2016). Given the observed decrease in hip flexion when wearing a softshell ankle brace, it is conceivable that wearing an ankle brace may alter the EMG activity of the muscles of the hip, altering important variables that may impact on vertical jump height performance.

Recent unpublished pilot data (Henderson & Sanzo, 2017; see Appendix J) has suggested that EMG activity of the lower extremity and lumbar spine is altered when wearing ankle braces during athletic performance tasks. Ten healthy, active individuals had his/her vertical jump height and cutting task time assessed under three conditions: no ankle brace, bilateral AE softshell ankle braces, and bilateral T1 ankle braces. During these tasks, the EMG of the PL, LG, BF, RF, GM, and ES was also collected. No differences were observed between conditions with respect to the time to complete the lateral cutting task, although a significant decrease in RF EMG activity was present during the deceleration phase of the cut. Similar to the findings reported by Smith et al. (2016), compared to wearing no ankle braces, a nonsignificant reduction of 1.87 cm in vertical jump height was observed when wearing the T1 ankle braces. Accompanying this reduction was a nonsignificant decrease in RF, GM, and ES EMG activity when wearing the T1 ankle braces. As such, it is conceivable that this reduction in RF, GM, and ES EMG activity may be contributing to the reduction observed in the vertical jump height. Because of the low sample size and lack of kinematic data, it is difficult to determine the degree to which this reduction in EMG activity may have affected vertical jump height. Irrespective of the mechanism, however, reductions in vertical jump height when wearing ankle braces have been demonstrated in the available literature.

Ankle Bracing and Athletic Performance

Due to their restrictive properties, many athletes subjectively perceive that ankle braces can reduce athletic performance (Rosenbaum et al., 2005). Given the importance of performance in high level sport, if wearing ankle braces were to negatively affect performance, this may reduce adherence to a prescribed ankle brace. Over the last few decades, a moderate amount of studies objectively investigated the effects of ankle bracing on athletic performance measures, such as vertical jump height and agility. Despite this, a consensus on whether or not ankle braces affect athletic performance has not been reached (Cordova, Scott, Ingersoll, & LeBlanc, 2005), with many studies presenting different conclusions. Reasons for this may include methodological differences between studies, as well as the brace models that were investigated.

To gauge the effect that ankle braces may have on athletic performance, several studies have used vertical jump height and time to complete an agility task as dependant variables. Henderson et al. (2016) assessed vertical jump height and time to complete the T-test Agility Test in 14 varsity jumping sport athletes when wearing the AE ankle brace and T1 semi-rigid ankle brace bilaterally. As measured with a Vertec™ device, countermovement vertical jump height was significantly decreased by 2.35 cm when wearing the T1 ankle braces, compared to the no ankle brace control condition. Similar effects were observed when wearing the AE ankle braces, although this reduction did not reach significance. Time to complete the T-test Agility Test was not significantly different between conditions.

Parsley et al. (2013) observed similar changes in countermovement vertical jump height when wearing the prototype Seattle Ankle Orthosis prototype external ankle brace, AE ankle brace, and Aircast Airsport™ semi-rigid ankle braces on both ankles. Twenty four physically active males experienced a significant decrease of 1.3-1.8 cm in all of the brace conditions, compared to when not wearing an ankle brace. As with Henderson et al. (2016), agility time was not affected, although the Modified Southeast Missouri Agility Run was used rather than the T-test Agility Test. Based on the results of these studies, it appeared that countermovement vertical jump height was impaired when wearing the semi-rigid and softshell ankle brace, while time to complete the agility task was unaffected when wearing the softshell or semi-rigid ankle braces. Nonetheless, research employing different methodologies and ankle braces presents evidence contrary to this conclusion.

As part of their investigation on the effects of ankle braces and ankle tape on athletic performance, Ambegaonkar et al. (2011) employed the Sargent Chalk Jump Test and Right-Boomerang Run Test to measure countermovement vertical jump height and agility time, respectively. Ten healthy participants performed each test under four conditions: no ankle braces, standard ankle athletic taping (basketweave), Swede-o® Ankle Lok™ softshell ankle braces, and Aircast Airstirrup™ semi-rigid ankle braces worn bilaterally. Vertical jump height was unaffected by wearing ankle braces or tape; however, agility time increased significantly by 0.59 seconds when wearing the Aircast Airstirrup™ ankle braces. These results suggested that vertical jump height was unaffected by the softshell or semi-rigid ankle braces, but agility performance was negatively affected; the opposite findings of Henderson et al. (2016) and Parsley et al. (2013). Further complicating matters, some studies have found no effect of ankle brace on performance.

Verbruggee (1996) investigated the effects of standard athletic taping applied to the ankle compared to the Aircast Airstirrup™ semi-rigid ankle brace on vertical jump height, agility time, and sprint time. Twenty six male collegiate athletes completed the Chalk Vertical Jump Test, Barrow and McGee Agility Run, and the 40 yard dash under three conditions in a random order: no ankle support, standard athletic taping (modified Gibney technique), and Aircast Airstirrup™ ankle brace worn on both ankles. No significant difference was observed between conditions with respect to agility time or sprint speed. There was an average decrease in vertical jump height of 1.52 cm when wearing the semi-rigid ankle brace compared to the no brace control condition, although this difference did not reach significance.

Similarly, insignificant reductions in vertical jump height were observed by MacKean et al. (1995) and Macpherson, Sitler, Kimura, and Horodyski (1995). MacKean et al. (1995) had 11 female basketball players perform a Wall Vertical Jump Test, jump shot, sprint drill, and treadmill run. In a randomized order, participants completed these tasks when wearing no ankle support, the T1 and Aircast Airstirrup™ semi-rigid ankle braces, Swede-o® softshell ankle brace, and standard athletic tape. Compared to wearing no ankle support, wearing the T1 and Aircast Airstirrup™ resulted in a reduced vertical jump height by approximately 2 cm. Additionally, sprint times were slightly slower in the ankle support conditions. Performing the treadmill run when wearing the Aircast Airstirrup™ ankle braces also slightly, but significantly increased energy expenditure and oxygen consumption. Macpherson et al. (1995) conducted a comparable study with 25 male high school varsity football players. When wearing the Aircast Airstirrup™ ankle brace or Donjoy Rocket Sock softshell ankle brace bilaterally during a Chalk Vertical Jump Test, sprint, and shuttle run, no significant differences were observed compared to the no brace control condition. The Aircast Airstirrup™ ankle braces, however, produced mean values that were 2.64 cm less than the no ankle brace control condition. Although these studies observed differences that did not reach significance, given the effects that even small decreases in performance can have in competitive sport, these decreases cannot be ignored. More recent literature, however, presents little differences between ankle brace and no ankle brace conditions.

Leonard and Rotay (2014) compared the effects of standard ankle athletic taping and generic softshell ankle braces on countermovement vertical jump height, power, and agility. In 10 athletic and nine non-athletic individuals, the effects on countermovement vertical jump height was similar across conditions, as measured by a Vertec™ device. Likewise, power generation and time to complete the Illinois Agility Test were not affected by wearing bilateral ankle taping or bracing. It is important to note that this study did not incorporate a semi-rigid ankle brace, which based on Henderson et al. (2016), MacKean et al. (1995), Parsley et al (2013)., and Verbrugee et al. (1996) appeared to affect vertical jump height more than softshell ankle braces. Furthermore, the brand of softshell ankle brace was not indicated, making it difficult to make comparisons to other studies based on the characteristics of the ankle brace. The results of this study are in line, however, with literature that assessed the effect of ankle braces on athletic performance in persons with CAI and a history of ankle sprains.

To assess the effects of ankle braces on sports performance, Rosebaum et al. (2005) conducted a comprehensive study of 10 different ankle braces inducing both objective and subjective measures. Thirty four participants, each self-reporting CAI, completed a 10 second obstacle course involving four tasks: maximum vertical jump, single leg-inclined hopping, sprinting, and a sidestep cutting task. Participants performed the obstacle course wearing four different softshell ankle brace models, four different semi-rigid ankle brace models, one rigid ankle brace model, and without ankle braces in a randomized order. Wearing no brace produced the highest vertical jump height, although not by a significant margin; both bilateral softshell and semi-rigid ankle braces produced similar values for all tasks when evaluated separately. With respect to the subjective measures, participants generally preferred the softshell ankle braces from a comfort and performance perspective, while participants preferred the semi-rigid ankle braces from an ease of use perspective. Irrespective of these measures, participants reported feeling similar levels of stability when wearing semi-rigid and softshell ankle braces, suggesting that any kind of support may be subjectively beneficial for persons with CAI during athletic performance.

Wiley and Nigg (1996) also examined participants with unstable ankles in their investigation of the effects of the Melloeloc semi-rigid ankle brace on athletic performance measures and ankle ROM. Both athletic performance and ROM were assessed before and after a standardized aerobic exercise session. For the athletic performance measures, 12 adolescent participants with a positive Anterior Drawer Test of the ankle and previous history of ankle sprain performed a vertical jump and a figure eight running test under two conditions: no ankle brace and wearing bilateral Melloeloc ankle braces. When participants performed a vertical jump with a five step, running approach, vertical jump height was similar when wearing no ankle braces and when wearing the Malleoloc ankle braces. This remained true before and after the exercise session. Additionally, no significant differences were observed in the time to complete the figure eight running task between the no ankle brace and Malleoloc ankle brace conditions, before and after the exercise session. There was, however, a small improvement in time to complete the figure eight running test when wearing the Malleoloc ankle braces post exercise session, as compared to before the exercise session. As participants had clinically unstable ankles, this suggested that the Malleoloc semi-rigid ankle brace may help maintain subjective feelings of stability as fatigue sets in, encouraging participants to maintain intensity. Conversely, this may be the result of participants becoming accustomed to the ankle braces.

The only study that has investigated both immediate and short term (1 week) effects of ankle braces on athletic performance measures was conducted by Pienkowski, Shapiro, Caborn, and Stayton (1995). Twelve adolescent male basketball players performed a wall vertical jump, standing long jump, cone run, and shuttle run under four conditions: no brace, wearing the Swede-o® Universal softshell ankle braces, Kallassy softshell ankle braces, or Aircast Airstirrup™ semi-rigid ankle braces. At the beginning of the week, participants performed each task under the corresponding condition. Participants implemented the corresponding condition for all games and practices for one week, after which all tasks were assessed again. This was done for all four conditions (4 weeks total). No significant differences were seen between conditions for any of the athletic performance tasks. There was, however, a small improvement in all athletic performance measures after 1 week of use, which suggested that participants were adapting to the ankle braces. Again, this could suggest a compensatory strategy to counteract

ROM restriction at the ankle; however, a reduction in ROM does not appear to affect performance on an obstacle course.

Using the SEBT and a military style obstacle course, Newman, Croy, Hart, and Saliba (2012) evaluated the effects of the McDavid Ultralite 195 softshell ankle brace on performance in military cadets. Thirty-seven military cadets performed the Star Excursion Balance Test (SEBT) and military obstacle course when wearing no ankle braces, and when wearing unilateral and bilateral ankle braces. The obstacle course involved a variety of tasks including: balance, crawling, running, and climbing. Unilateral and bilateral ankle bracing produced significant differences in anterior and posterior reach distance. Despite this, the time to complete the obstacle course was not affected by either unilateral or bilateral ankle bracing. As there was a reduction in SEBT, this indicated that mobility and ROM was affected when wearing an ankle brace, although this reduction did not seem to manifest in a reduction on the time to complete a military obstacle course. Therefore, as unilateral and bilateral ankle bracing produced similar obstacle course results, this may suggest biomechanical adaptations to counteract reduced mobility in the ankle.

Purpose of Research

While ankle braces have been proven effective for reducing ankle injuries, there is still controversy surrounding the effects that they may have on other areas of the body. Although results have been inconsistent, studies have noted decreases in athletic performance and increases in non-ankle lower extremity injuries when wearing ankle braces. These results have been attributed to ankle braces unfavourably altering kinematics and kinetics (Smith et al., 2016). Little attention, however, has been given to the effects that ankle braces may have on the EMG activity of the lower extremity muscles, specifically muscles of the knee and hip joints. Given observed changes in knee flexion angles (Hodgson et al., 2005; Simpson et al., 2013) when wearing ankle braces, it is possible that EMG activity of the lower extremity may be similarly affected by wearing ankle braces. The implications of altered EMG activity when wearing ankle braces would be two-fold: 1) the ability to generate torque about the hip and extension at the knee are critical to vertical jump performance (Aragón-Vargas, & Gross, 1997), thus changes in EMG activity of the knee and hip musculature may affect vertical jump height; and 2) muscular activation of the lower extremity is important for attenuating GRFs (Devita & Skelly, 1992) and maintaining knee stability (Draganich, Jaeger, & Kralj, 1989). As such, altered EMG activity of the lower extremity when wearing ankle braces may affect the body's ability to attenuate GRFs and maintain knee stability, both of which are important for injury prevention. As ankle braces represent the most common method of ankle injury prevention (Stasinopoulos, 2004), it is essential that athletes and healthcare providers are aware of all possible effects of wearing braces, including those involving EMG activity of the lower extremity, forces, and athletic performance. Therefore, the purpose of this study was to determine if wearing semi-rigid and softshell ankle braces affected vertical jump height and the time to complete a lateral cutting task, as well as lower extremity EMG activity and kinetics during these measures.

Research Questions

The following questions were used to guide this study:

1) Is there a difference amongst ankle brace conditions in vertical jump height for the Vertical Jump Test?
2) Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the takeoff phase of the Vertical Jump Test?

3) Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the landing phase of the Vertical Jump Test?
4) Is there a difference amongst ankle brace conditions in time to complete the cutting task?
5) Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the deceleration phase of the cutting task?
6) Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the propulsive phase of the cutting task?

For questions one and four, a one-way ANOVA for repeated measures was conducted to compare the independent variable (brace condition) on vertical jump height, as well as time to complete the lateral cutting task. For questions two and three, a one-way ANOVA for repeated measures was conducted to compare the independent variable (brace condition) on mean EMG activity of the PL, LG, BF, RF, and GM muscles, peak vertical, mediolateral, and anteroposterior GRFs, and total vertical impulse. For questions five and six, a one-way ANOVA for repeated measures was conducted to compare the independent variable (brace condition) on mean EMG activity of the PL, LG, BF, RF, and GM muscles, peak vertical, mediolateral, and anteroposterior GRFs, and total vertical, anteroposterior, and mediolateral impulse.

Chapter Two: Methodology

A limited number of studies have examined the effects of ankle braces on lower extremity muscle activity, kinetics, and performance during athletic performance measures. As such, this study was designed and conducted with the intent of filling this gap in the literature.

Participant Inclusion and Exclusion Criteria

Participants were included into the study if they: 1) were male or female, physically active individuals; 2) participated in at least 150 minutes of moderate to vigorous aerobic activity each week, 3) had previous experience with “jumping” and/or “cutting” sports (i.e., basketball, volleyball, soccer) and; 3) were between the ages of 18-30 years. This specific population was selected to represent a physically active group that may/may not have experience using ankle braces and to maximize recruitment potential.

Participants were excluded from this study if they: 1) suffered from a diagnosed or self reported ankle injury over the last 6 months (e.g., sprain, fracture, tendonitis); 2) were currently suffering from an acute and/or chronic lower extremity or lumbar spine injury (i.e., strain or sprain) that precluded them from participating in jumping or running activities; 3) had undergone any lower extremity or lumbar spine surgical procedure in the last six months; and 4) were allergic or sensitive to adhesive tape or any of the material present in the AE and T1 ankle braces (i.e., Velcro or plastic).

Research Recruitment Procedures

Participants were recruited via purposive and convenience sampling, as the targeted population was physically active individuals in Thunder Bay and Northwestern Ontario. A total of 42 participants were recruited into the study. Based on pilot data and a priori analysis, this number was sufficient to detect a medium to large effect size with 80% power at α=.05 (two-tailed) in the Vertical Jump Test and cutting task performance measures. Participants were recruited using recruitment posters containing an overview of the study and contact information for the researcher (see Appendix A). These posters were posted in visible, high traffic areas throughout the Sanders Building at Lakehead University, in addition to the School of Kinesiology and Lakehead University social media accounts. Participants were also recruited through word of mouth, where he/she was encouraged to contact the researcher if he/she was interested in participating in the study. When a potential participant contacted the researcher and expressed interest in participating in the study, he/she was provided with information pertaining to the study including: an information letter (see Appendix B), informed consent form (see Appendix C), Physical Activity Readiness Questionnaire (PAR-Q) form (see Appendix D), and Functional Ankle Ability Measure (FAAM) outcome questionnaire (see Appendix E) for his/her perusal. The researcher also provided answers to any questions that the participant had about the study. The researcher then set up testing sessions with the participant at mutually agreeable times.

Screening Measures

Physical Activity Readiness Questionnaire. The Physical Activity Readiness Questionnaire is a pre-screening form developed by the Canadian Society for Exercise Physiology (CSEP) to identify any medical conditions that may make physical activity unsafe for an individual (CSEP, 2013). The PAR-Q was completed by participants prior to participating in any physical components of this study. Although no participants did not pass the PAR-Q pre-screen, if a participant answered yes to any of the questions on the PAR-Q, he/she would have been deemed ineligible to participate in the study and would have been encouraged to consult with his/her healthcare provider as soon as possible.

Functional Ankle Ability Measure. The Functional Ankle Ability Measure (FAAM) is a self-report measure of ankle function designed to quantify functional limitations of the ankle (Eachaute, Vaes, van Aerschot, Asman, & Duquet, 2007). As such, it was used demographically to determine the level of ankle function in each participant. The FAAM is comprised of two subscales: an activities of daily living (ADL) subscale and a sports subscale. The ADL subscale is made up of 21 items, while the sports subscale is comprised of eight items. It is scored on a 5-point Likert scale (0-4), with 0 representing *no difficulty* and 4 representing *unable to do*. The ADL and sport subscale are scored separately, with a maximum possible score of 84 for the ADL subscale, and 34 for the sport subscale. The ADL and sports subscales have both been shown to be reliable, with intraclass correlation coefficients (ICC) of .89 for the ADL and .87 for the sports subscale (Eachaute et al., 2007). Furthermore, the ADL and sports subscales display some evidence of moderately high construct validity with the SF-36 physical subscale ($r=.84$; $r=.78$; Eachaute et al., 2007) As such, ankles with minimal functional limitations will consistently score higher than ankles with functional limitations. Based on previous research and for demographic purposes, participants scoring less than 85% on the sports subscale were considered to have diminished ankle function (Barlow et al., 2015).

Instrumentation

Electromyography. Surface electromyography (sEMG) was used as a non-invasive procedure to measure neuromuscular activity (Konrad, 2005). Surface electromyography has demonstrated high reliability (ICCs greater than .80) when measuring quadriceps and hamstring muscle activity during dynamic tasks (Fauth, Petushek, Feldmann, Hsu, & Garceau, 2010). A Delsys Trigno™ Wireless EMG system and Trigno™ IM sensors were used for this study. The Delsys Trigno™ EMG system and accompanying sensors are capable of collecting 16 EMG channels simultaneously, with a transmitting range of 20 metres (Delsys Inc., 2012). For the purposes of this study, the Delsys Trigno™ Wireless EMG system was connected to an interface board, which was connected to a PowerLab (16/30) data acquisition unit. Electromyographic activity of the PL, LG, BF, RF, and GM muscles, measured in millivolts (mV) was collected.

Advanced Medical Technologies Incorporated force platform. A force platform was used to measure kinetic variables. The AMTI force platform has demonstrated excellent reliability (ICC=.94) when assessing vertical GRF during a vertical jump (Cordova & Armstrong, 1996). For the purposes of this study, designated channels for vertical, anteroposterior, and mediolateral GRFs were connected to a PowerLab (16/30) data acquisition unit. Vertical, anteroposterior, and mediolateral forces were measured in Volts (V). These values were then converted to Newtons (N); an external load cell was regularly used to calibrate the force platform and determine the appropriate V to N ratio in LabChart®. See figure 6 for force plate and EMG set-up.

LabChart® software. For this study, EMG and GRF data were simultaneously collected in real time by LabChart© software.. For the purposes of this study, three channels (Fz, Fy, and Fx) collecting GRF data from the AMTI force platform, and five channels (PL, LG, BF, RF, GM) collecting EMG data from the Delsys Trigno™ Wireless EMG system were fed into the PowerLab unit for real time synchronization. Labchart® software was used to rectify, filter, and extract all EMG and force platform data.

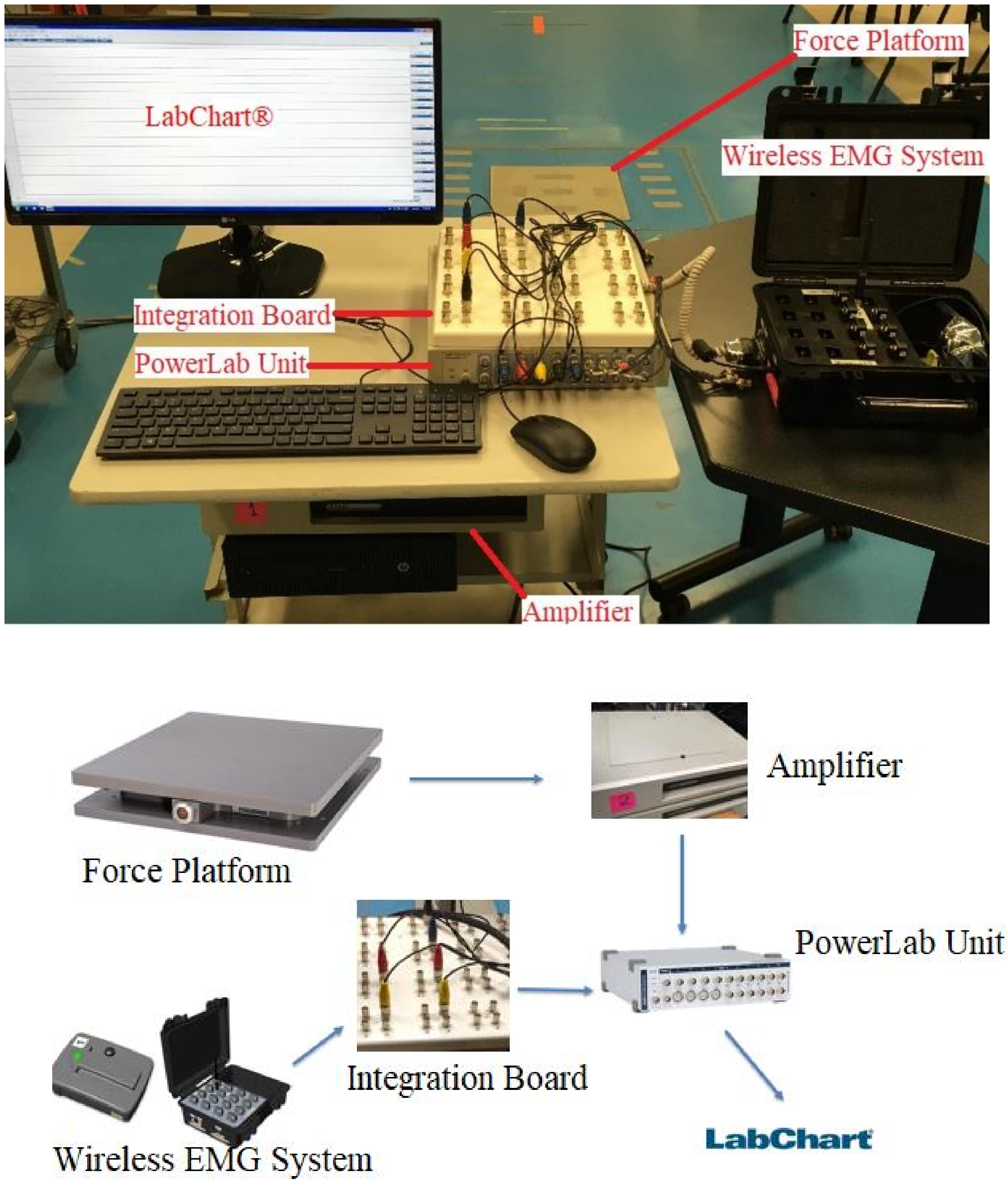


Figure 6. Instrumentation set-up. This figure visually represents the configuration of all instruments. The force platform was connected to the amplifier, of which three channels were fed to the PowerLab Unit. The wireless EMG system was connected to the PowerLab unit via an integration board, with five channels being fed into the unit. LabChart® data collected data from both the force platform and wireless EMG system in real time.

Brower timing gates. To measure cutting task time, Brower infrared timing gates were used to create a start and finish line for the cutting task. Timing gates create a starting line using an infrared signal to connect multiple sensors. When this signal was broken by a participant for the first time, the timer started. When the participant crossed the line again, the time between infrared disruptions was recorded as the cutting task time. Timing gates have been shown to have a high intra trial reliability, with a coefficient of variation (CV) of 0.69 to 1.2% when measuring 10 metre sprint speed (Cronin & Templeton, 2008).

Vertical Jump Test. Vertical jump height is a widely-used measure of lower body power and is pertinent to anaerobic and jumping sports (Aragón-Vargas, 2000; CSEP, 2013). Several methods have been developed over the last decade; however, the CSEP procedures for vertical jump was used in this study (see Appendix G) with the addition of a Vertec™ device (see Appendix G). This is due to the relative ease of use, standardization of procedures, availability of normative data, and low use of special equipment (CSEP, 2013). Furthermore, the Vertec™ device has demonstrated good intrasession and intersession reliability for males (ICC=.94; ICC=.90) and females (ICC=.87; ICC=.80) when measuring a countermovement vertical jump height (Nuzzo, Anning, & Scharfenberg, 2011). For the purposes of this study, vertical jump height (in) was recorded and converted to cm.

Cutting task. For the purposes of this study, a cutting task was used to assess lateral cutting (see Appendix H). Due to laboratory constraints, the cutting task was adapted from the T-Test Agility Test as it incorporates movement in both the sagittal and frontal planes (Pauole, Madole, Garhammer, Lacourse, & Rozenek, 2000). A similar modified test, however, was used by Klem et al. (2016). This task was developed specifically for this study and, as a result, no psychometric properties are available with respect to its ability to reliably measure agility, although the full T-Test Agility Test has demonstrated excellent reliability (ICC=.98; Pauole et al., 2000). For the purposes of this study, time (sec) to complete the cutting task was recorded.

Procedures

Two testing sessions were required to collect data. The first session lasted approximately 60 minutes and the second session lasted approximately 45 minutes. Testing took place in room SB-1028 in the Sanders Building at Lakehead University. At the first testing session, the researcher provided an overview of the study, and completed the informed consent process with the participant, answering any questions that he/she had. If the participant chose to partake in the study, he/she signed the consent form and filled out the FAAM and PAR-Q. Once these forms were completed, descriptive, and background information (see Appendix F) were recorded including: age (years), height (centimetres), mass (kilograms), and experience wearing an ankle brace (yes/no).

Following the collection of descriptive and background data, the participant was assigned to one of six intervention sequences, based upon the order in which he/she was recruited, to counterbalance the intervention order. The athletic performance measures of the study were performed under each of three possible conditions: no brace control, bilateral AE softshell ankle braces, and bilateral T1 ankle braces. As such, six possible intervention orders and groups were possible (see Table 1).

Table 1.
Intervention sequences.

Order	Sequence
1	no brace, AE, T1
2	no brace, T1, AE
3	AE, no brace, T1
4	AE, T1, no brace
5	T1, no brace, AE
6	T1, AE, no brace

Following group assignment and collection of baseline information, the participant completed a 5 minute warmup on a cycle ergometer at an intensity of 10-12 on the Borg Scale of

Perceived Exertion (CSEP, 2013). A visual representation of the Borg Scale was shown to the participant to help he/she determine the appropriate intensity (see Appendix I).

Following the warmup, six IM wireless electrodes of a Delsys Trigno™ Wireless EMG system were applied by the researcher to the skin of the dominant leg of the participant, defined as the foot with which the participant would kick a ball. Before applying the electrodes, the researcher asked the participant for consent to place the electrodes on his/her skin. After verbal consent was obtained, the researcher prepared the electrode sites by cleaning the area with isopropyl alcohol and shaving the area, if necessary. This removed any dry skin and hair that may have affected electrode adherence and signal attenuation (Delsys Inc., 2012). Following the application of each electrode, an MVC was performed and recorded for the corresponding muscle. Electrode application and MVCs were performed systematically in the following order: PL, LG, BF, GM, RF.

The first electrode was applied to the PL muscle. To landmark the position for placement of the electrode, the participant was positioned in supine lying, with the knee slightly flexed and medially rotated. The electrode was placed 25% of the distance from the head of the fibula to the lateral malleolus (Figure 7; SENIAM, 2017). The participant then completed an MVC for the PL muscle; from a supine lying position, with the arms supporting the trunk and leg in slight flexion, the participant everted and plantarflexed the foot while the researcher provided manual resistance for 3 seconds (SENIAM, 2017).

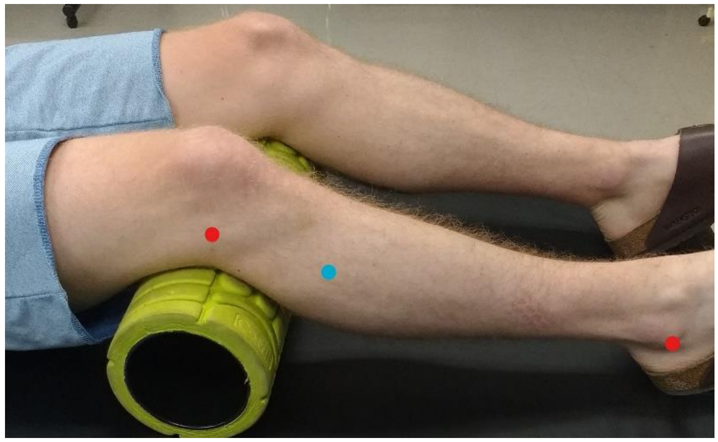


Figure 7. Electrode location one. This figure displays the location for the PL muscle. The red dots represent anatomical landmarks. The blue dot represents the electrode location.

The second electrode was applied to the LG muscle. To landmark the position for placement of the electrode, the participant was positioned in prone lying with the knee slightly flexed and foot extending over the end of the table. The electrode was placed 33% of the distance from the head of the fibula to the calcaneus (Figure 8; SENIAM, 2017). The participant then completed an MVC for the LG muscle; from a prone lying position with the knee in full extension, the participant plantarflexed the foot while the researcher provided manual resistance for 3 seconds (Halak & Ginn, 2012; Hsu, Krishnamoorthy, & Scholz, 2009).

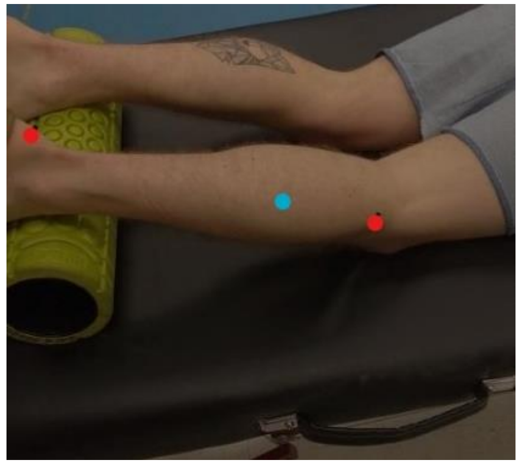


Figure 8. Electrode location two. This figure displays the location for the LG muscle. The red dots represent anatomical landmarks. The blue dot represents the electrode location.

The third electrode was applied to the BF muscle. To landmark the position for placement of the electrode, the participant was positioned in prone lying, with the knee in a flexed position that was less than 90° and in minimal lateral rotation. The electrode was placed halfway between the ischial tuberosity and the lateral epicondyle of the tibia (Figure 9; SENIAM, 2017). The participant then completed an MVC for the BF muscle; in prone lying and slight lateral rotation, the participant flexed the knee (from 60°) while the researcher provided manual resistance for 3 seconds (Halak & Ginn, 2012; Hsu et al., 2009).

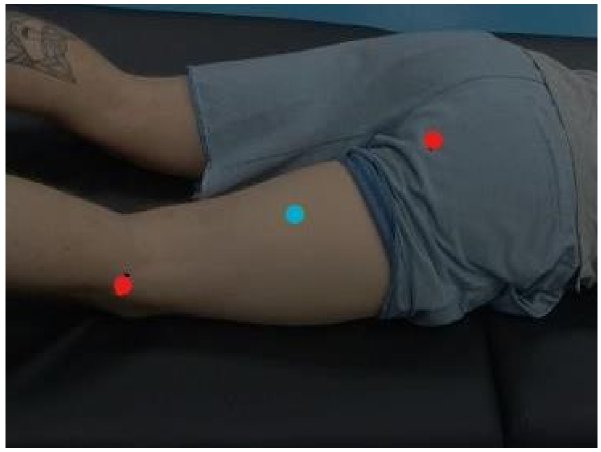


Figure 9. Electrode location three. This figure displays the location for the BF muscle. The red dots represent anatomical landmarks. The blue dot represents the electrode location.

The fourth electrode was applied to the GM muscle. To landmark the position for placement of the electrode, the participant was positioned in side lying. The electrode was placed halfway between the iliac crest and the greater trochanter (Figure 10; SENIAM, 2017). The participant then completed an MVC for the GM muscle; in a side lying position, the participant abducted the hip 25° while the researcher provided manual resistance for 3 seconds (Bolga & Uhl, 2007; Halak & Ginn, 2012)

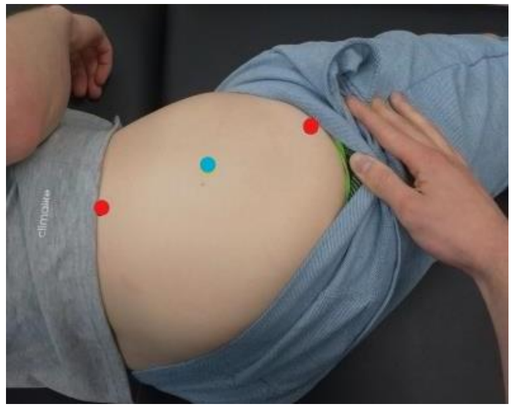


Figure 10. Electrode location four. This figure displays the location for the GM muscle. The red dots represent anatomical landmarks. The blue dot represents the electrode location.

The final electrode was applied to the RF muscle. To landmark the position for placement of the electrode, the participant was positioned in supine lying. With the upper body supported by the participant's arms and the hip in slight flexion, the electrode was placed halfway between the anterior superior iliac spine and the superior border of the patella (Figure 11; SENIAM, 2017). The participant then completed an MVC for the RF muscle; in a seated position, with the hips flexed to 90°, the participant extended the knee (from 60°) while the researcher provided manual resistance for 3 seconds (Halak & Ginn, 2012; Hsu et al., 2009)

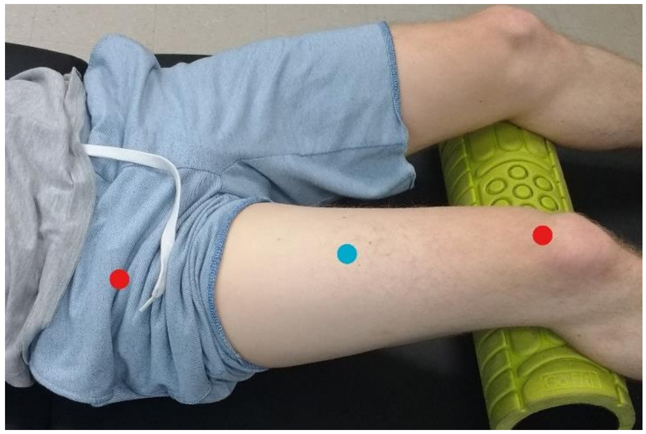


Figure 11. Electrode location five. This figure displays the location for the RF muscle. The red dots represent anatomical landmarks. The blue dot represents the electrode location.

After recording the EMG activity while completing the MVC, the participant applied the ankle braces (if necessary). Ankle braces were sized and applied by the participant based on the manufacturer's guidelines. These instructions were available in print form for the participant, and verbally stated to the participant by the researcher. A selection of ankle brace sizes was supplied by the researcher. The researcher supervised the application of all braces to ensure braces were applied properly. The participant was allowed to adjust the ankle braces during the testing session, if required, to maintain the manufacturer's described fit. Ankle braces were applied without modifying the participant's normal training shoe.

Following ankle brace application (if necessary), the researcher familiarized the participant with the Vertical Jump Test via verbal explanation and visual demonstration. To perform the Vertical Jump Test, the participant positioned the dominant foot over the AMTI force platform. While maintaining contact with the force platform, the participant adjusted his/her legs to a comfortable width. If necessary, the Vertec™ device was adjusted so that it was an appropriate distance from the participant without having it contact the force platform. The participant initiated the jump by flexing at the hip and knees and lowering into a 45° squatted position. While descending, the participant moved his/her arms in a counterweight fashion. In this position, the participant's form was visually evaluated by the researcher to ensure consistent form across trials and participants; if the participant's position was not acceptable, the researcher stopped the test. Participants held this 45° squatted position for 2 seconds, before jumping and touching the Vertec™ device as high as possible. At landing, the participant's weight was to be evenly distributed over both feet, with the dominant foot, in its entirety, contacting the force platform. A trial was considered successful if the participant paused for 2 seconds in the 45° squatted position and landed on two feet, with the dominant foot in contact with the force platform.

A maximum of five submaximal attempts were allowed for the participant to become comfortable with the test, ensure equipment was in working order, and allow the researcher to provide feedback regarding the participant's form. Once the participant was familiar with the test, he/she performed the first of three recorded trials. The researcher began collecting EMG and GRF data and notified the participant; when the participant was ready, he/she performed a maximal effort jump. The EMG and GRF data were collected during takeoff and landing, and ceased after the system weight normalized. If the researcher deemed the jump satisfactory, the height was recorded for the flight phase of the jump, in addition to EMG and GRF data. The participant then performed the test for two more recorded trials, spaced approximately 1 minute apart.

After completing the Vertical Jump Test under the first condition, a 5 minute rest period was provided. The purpose of this rest period was to allow the participant to transition to the next component of the testing protocol and allow for physical recovery. If the application of ankle braces was required, the participant did so during this time. Once the participant transitioned to the next condition, testing procedures mirrored that of the first condition, as described in detail previously. After the completion of the second set of trials, another 5 minute rest period took place, allowing the participant to transition to the last condition. Testing procedures followed the same procedures outlined for the previous two conditions.

Following completion of the Vertical Jump Test under all three conditions, the participant removed any ankle braces (if necessary) and electrodes. Once ankle braces and electrodes were removed, the participant completed a 5 minute cooldown on a cycle ergometer at a perceived exertion level of 10-12 on the Borg Scale. A visual aid was provided to the participant during this time to help he/she attain the desired intensity. The testing session concluded after completion of the 5 minute cooldown.

The second testing session was performed as soon as possible following the completion of the first session, with a minimum of 24 hours between sessions and a maximum of 10 days. All participants were asked to wear the same footwear that were worn for the first testing session. The second testing session began with the warmup, application of the electrodes, MVCs, and application of the appropriate braces (as per the pre-selected order). After the electrodes and ankle braces (if necessary) had been applied, the cutting task was explained to the participant (Figure 12). The participant began the test five metres away from the force platform, where the starting line and timing gates were set up. The participant positioned him/herself in an athletic position, with his/her dominant limb forward in a staggered stance. When the participant was ready to start the test, he/she crossed cross the timing gate, initializing the timing mechanism. The participant ran forwards toward the force platform; a cone was positioned directly in front of the force platform for the participant to touch. The participant touched the cone, simultaneously planting the dominant foot on the force platform. Immediately after touching the cone, the participant side shuffled to his/her left or right for three metres, based on the planting foot, towards another cone; the direction that the participant side shuffled was the same for all subsequent trials. The participant touched this cone and then side shuffled back to the first cone. After touching the first cone and force plate for the second time, the participant backpedaled towards the starting line. The test concluded once the participant crossed the timing gate at the start line.

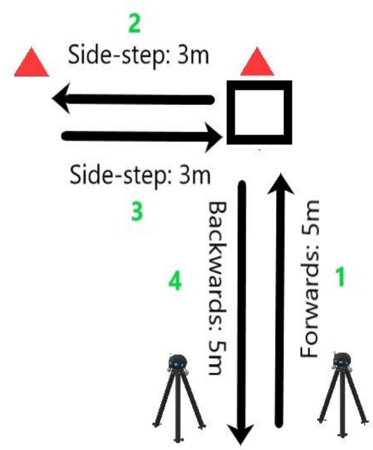


Figure 12. Cutting task. This figure displays the set up for the cutting task.

The participant was asked to complete at least two submaximal tests to familiarize him/herself with the cutting task, and to ensure that the equipment was functioning properly. Once the participant was familiar with the test, he/she performed the first of three recorded trials. The researcher began collecting EMG and GRF data and notified the participant; when the participant was ready, he/she initiated the test by crossing the timing gate. The EMG and GRF data were collected from the moment the researcher notified the participant, to when the participant crossed the timing gate at the finish. If the participant successfully contacted the force platform with the entire foot, EMG, GRFs, and time to complete the lateral cutting task was recorded. If the participant did not contact the force platform with the entire foot during the first contact, the test was stopped immediately, and no data were recorded. Following the completion of the first trial, the participant completed the task two more times, spaced approximately 1 minute apart.

After completing three recorded trials under the first condition, the participant transitioned to the next component of the testing protocol in the same manner as the first testing session. The second and third set of trials were completed in the same fashion as the first set of trials. Following completion of the cutting task under all three conditions, the participant removed any ankle braces (if necessary) and electrodes. Once ankle braces and electrodes were removed, the participant completed a 5 minute cooldown on a cycle ergometer at a perceived exertion level of 10-12 on the Borg Scale. A visual aid was provided to the participant during this time to help he/she attain the desired intensity. The testing session concluded after completion of the 5 minute cooldown.

Data Processing

Vertical Jump Test. The Vertical Jump Test was divided into three phases; takeoff, flight, and landing. For the purposes of EMG and GRF analysis, the takeoff and landing phases were examined. The takeoff phase (a) was defined as the time at which vertical GRF began to increase (greater than 5 N) from the stationary system weight, to the time that the system weight equalled 0 N (+/- 5 N). The landing phase (b) was defined as the time at which system weight increased from 0 N (greater than 5 N) to the peak vertical GRF value (+/- 5 N; see Figure 13). Force platform data were collected at 1000 samples per second.

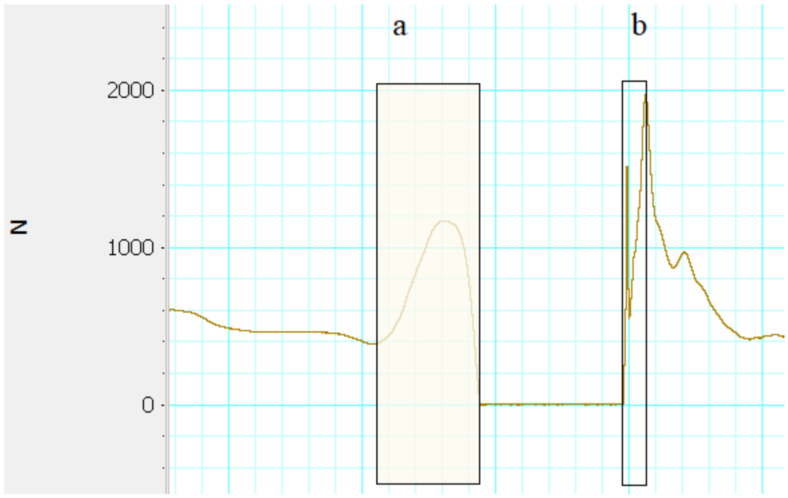


Figure 13. Vertical Jump Test takeoff and landing. This figure illustrates the take off and landing phase of the Vertical Jump Test, based on vertical GRF.

Lateral cutting task. For the purposes of EMG and GRF analysis, the cutting task was divided into two phases; deceleration and propulsive. The deceleration phase (1a) of the first platform was defined as the time at which anteroposterior GRF began to decrease (greater than 5 N) from 0 N to the minimum anteroposterior GRF value (+/- 5 N). The propulsive phase (1b) was defined as the time of peak anteroposterior GRF (+/- 5 N) to the time that system weight equalled 0 N (+/- 5 N; Figure 14. Force platform data were collected at 1000 samples per second.

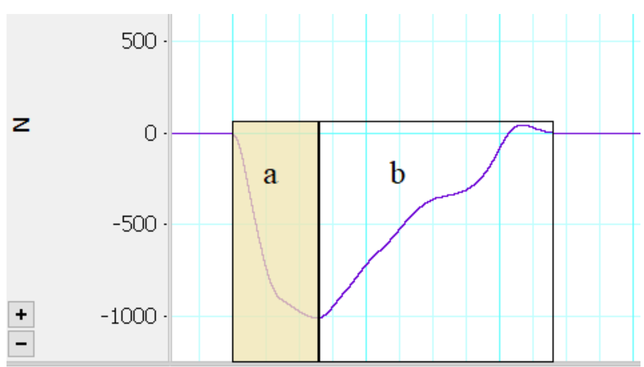


Figure 14. Lateral cutting task first contact anteroposterior GRF. This figure illustrates the deceleration phase (1a) and propulsive phase (1b) of the first contact with the force platform, based on anteroposterior GRF.

Electromyography data. Raw EMG data were bandpass filtered with a low and high cut-off frequency of 500 Hz and 10 Hz, respectively. This was done to minimize movement artifact associated with dynamic movements, and based on recommendations by Stegeman and Hermens (1998) and Merletti, Farina, Hermens, Freriks, and Harlaar (1999) for the processing of sEMG for dynamic movements. Following bandpass filtering, the data were full wave rectified to create a linear envelope (Kamen & Gabriel, 2010).

Preliminary Data Analysis

All electromyographic and force data were extracted from LabChart© and transferred to a Microsoft Excel spreadsheet for the purposes of data management and preliminary analysis.

Electromyography data. Mean EMG activity (mV) for the 3 seconds MVC was first calculated. Mean EMG activity (mV) for each muscle during each phase of the Vertical Jump Test and cutting task was then calculated and averaged across the three recorded trials after being checked for reliability using a two-way, mixed effects intraclass correlation (Shrout & Fleiss, 1979; Koo & Li, 2016). The average of the three trials for each condition and phase was then expressed as a percentage of MVC (%MVC; Halaki & Ginn, 2012).

Force platform data. Force platform data for each phase of the Vertical Jump Test and cutting task were averaged across the three recorded trials after being checked for reliability using a two-way, mixed effects intraclass correlation (Shrout & Fleiss, 1979; Koo & Li, 2016). The average of the three trials for each condition and phase was then expressed as a percentage of bodyweight (%BW).

Performance data. Vertical jump height (cm) was averaged across the three recorded trials for each condition after being checked for reliability using a two-way, mixed effects intraclass correlation (Shrout & Fleiss, 1979; Koo & Li, 2016). Time to complete the lateral cutting task (sec) was averaged across the three trials for each condition after being checked for reliability using a two-way, mixed effects intraclass correlation (Shrout & Fleiss, 1979; Koo & Li, 2016).

Statistical Analysis

Following preliminary analysis, data were transferred into an IBM SPSS 24 data file for further analysis. For each research question, a one-way ANOVA for repeated measures was conducted to compare the independent variable (brace condition) on each dependant variable. To minimize type I error, a Bonferroni adjustment ($p < .05/5$) was applied to the analysis of EMG and GRF measures for the vertical jump takeoff and landing, resulting in an alpha level of $p < .01$. Similarly, a Bonferroni adjustment ($p < .05/5$) was applied to the analysis of EMG and GRF measures for the deceleration and propulsive phases of the cutting task, resulting in an alpha level of $p < .01$. A Bonferroni adjustment($p < .05/3$) was also applied to analysis of impulse measures, resulting in an alpha level of $p < .017$. For vertical jump height and time to complete the cutting task, alpha level was set at $p < .05$.

The independent variable consisted of three levels: no brace control, AE ankle braces, and T1 ankle braces (T1). For the Vertical Jump Test dependent variables included vertical jump height, mean EMG activity of the PL, LG, BF, RF, and GM muscles, peak vertical, mediolateral, and anteroposterior GRFs, and total vertical impulse during takeoff and landing. For the cutting task, dependent variables included time to complete the cutting task, mean EMG activity of the PL, LG, BF, RF, and GM muscles, peak vertical, mediolateral, and anteroposterior GRFs, and total vertical, anteroposterior, and mediolateral impulse during the deceleration and propulsive phase.

For all variables, boxplots were visually inspected to determine the presence of outliers; all analyses were performed with and without outliers to determine if they influenced the results. The data were then checked for normality using the Shapiro-Wilk test for normality. If data were not normally distributed, the Friedman’s test was conducted, in addition to a one-way ANOVA for repeated measures. Data were assessed for sphericity using Mauchly's Test of Sphericity. If the data violated the assumption of sphericity, a Greenhouse-Geisser correction was applied.

Chapter Three: Vertical Jump Test Results

The results of this study provide evidence that performance, measures of lower extremity EMG muscle activity, performance (e.g., vertical jump height) and GRFs may be affected during a Vertical Jump Test and cutting task when wearing softshell and semi-rigid ankle braces.

Demographics

A total of 42 participants were recruited into the study. Demographic information of all participants is presented in Table 1.

Table 2.
Participant demographic information.

Sex	23 male, 19 female
Height (cm)	174.47 +/- 8.06
Mass (kg)	77.8 +/- 13.31
Age (years)	22.16 +/- 1.88
Dominant Foot	37 right, 5 left
Experience	22 yes, 20 no
Ankle Function	42 normal, 0 diminished

Missing Data

Due to equipment malfunction (electrode failure, loss of electrode adherence, force platform failure), usable data for each variable varies from n = 34 to n = 41. In cases where only two out of three trials were considered usable due to equipment malfunction, the mean of the two trials was used in place of a third trial when calculating the average of the three trials.

Takeoff and Flight Phase

Reliability. Intraclass correlations for all dependent variables are presented in Table 3.

Table 3.
Intraclass correlations for Vertical Jump Test performance and takeoff dependent variable across three trials.

Dependent Variable	**No Brace**	**ASO® EVO®**	**Active Ankle T1™**
PERFORMANCE			
Vertical Jump Height (n = 41)	.961	.976	.998
EMG			
PL Mean EMG (n = 33)	.739	.877	.980
LG Mean EMG (n = 35)	.844	.859	.897
BF Mean EMG (n = 40)	.761	.870	.760
RF Mean EMG (n = 38)	.814	.927	.860
GM Mean EMG (n = 34)	.877	.826	.939
GROUND REACTION FORCE			
Peak Vertical GRF (F_z; n = 40)	.975	.984	.975
Peak Posterior GRF (F_y; n = 40)	.849	.820	.874
Peak Anterior GRF (F_y; n = 40)	.875	.896	.845
Peak Lateral GRF (F_x; n = 40)	.927	.863	.843
Peak Medial GRF (F_x; n = 40)	.917	.790	.761
Total Vertical Impulse (n = 40)	.897	.892	.911

Descriptive statistics. Descriptive statistics for all dependent variables during the takeoff phase of the Vertical Jump Test are presented in Table 4.

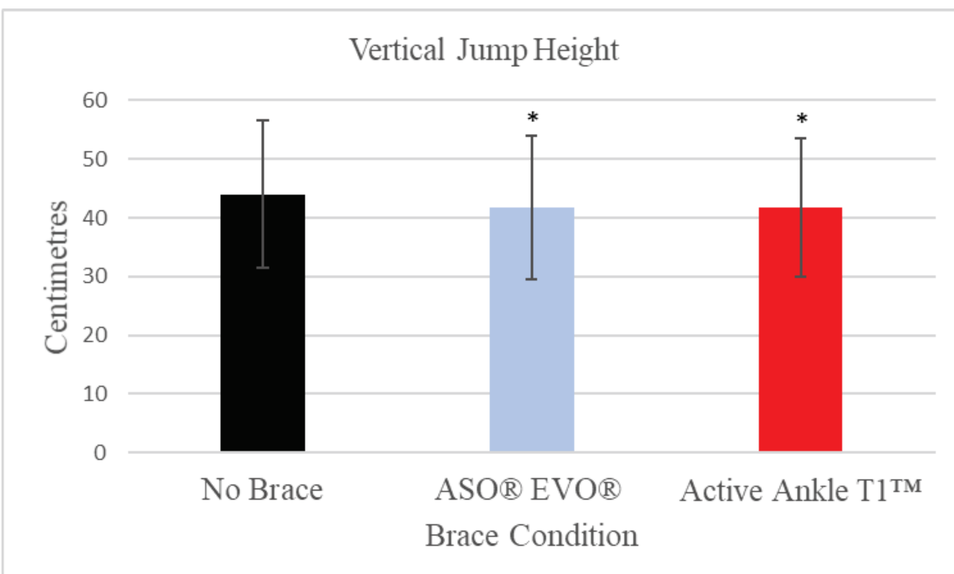
Table 4.
Electromyography and kinetic descriptive statistics for the takeoff phase of the Vertical Jump Test.

Dependant Variable	No Brace Mean	SD	n	ASO® EVO® Mean	SD	n	Active Ankle T1™ Mean	SD	n
EMG (%MVC)									
PL Mean EMG	69.22	15.21	33	65.42	14.71	33	66.93	15.61	33
LG Mean EMG	73.45	34.83	35	70.67	35.99	35	66.11	36.47	35
BF Mean EMG	30.53	14.54	40	33.05	20.22	40	32.89	19.5	40
RF Mean EMG	98.81	38.67	38	91.65	34.66	38	98.58	38.59	38
GM Mean EMG	55.33	34.07	34	53.98	35.07	34	55.35	34.13	34
GRF (%BW)									
Peak Vertical GRF (F_z)	115.3	12.57	40	115.77	12.85	40	115.41	13.33	40
Peak Posterior GRF (F_y)	10.31	5.69	40	10.71	5.37	40	10.68	6.05	40
Peak Anterior GRF (F_y)	13.51	5.88	40	13.21	5.87	40	13.25	6.06	40
Peak Lateral GRF (F_x)	10.31	5.69	40	16.95	4.95	40	17.07	5.18	40
Peak Medial GRF (F_x)	0.198	2.61	40	2.21	2.8	40	2.58	2.96	40
Total Vertical Impulse (N·s)	286.1	70.2	40	284.62	68.7	40	288.87	65.17	40

Inferential statistics.

Question one: Is there a difference amongst ankle brace conditions in vertical jump height for the Vertical Jump Test?

As assessed by boxplot inspection, there was one outlier in the T1 condition. This outlier was considered to be a true value and, therefore, was included in the analysis. Furthermore, the analysis was conducted with and without outlier data and produced similar results. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). A one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on vertical jump height. There was a significant change in vertical jump height (n = 41), $F(2, 80) = 15.796$, $p < .001$, $\eta p^2 = .283$. Mean vertical jump height decreased from 43.94 +/- 12.55 cm when wearing no brace to 41.84 +/- 12.22 cm and 41.81 +/- 11.81 cm when wearing the AE and T1 ankle braces, respectively. Bonferroni pairwise comparisons analysis revealed a significant decrease in vertical jump height when wearing the AE ankle brace compared to no brace (2.09 (95% CI, 0.9 to 3.28) cm, $p < .001$). Additionally, there was a significant decrease in vertical jump height when wearing the T1 ankle brace compared to no brace (2.12 (95% CI, 1.058 to 3.193) cm, $p < .001$). There was no significant difference in vertical jump height between AE and T1 ankle brace conditions (.03 (95% CI, -.948 to 1.01) cm, $p > .05$; see Figure 15)



*Significantly different from no brace condition ($p < .05$)

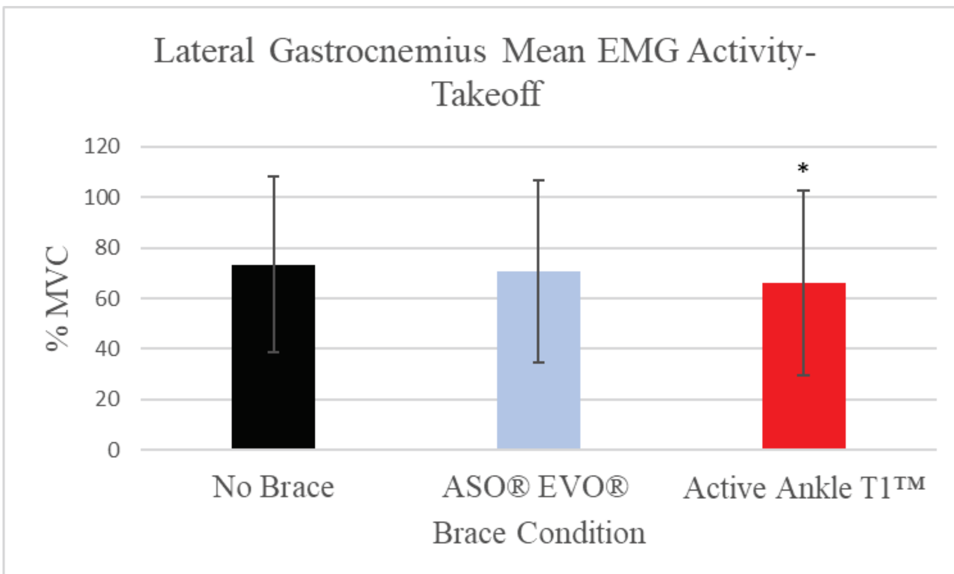
Figure 15. Differences in vertical jump height activity across brace conditions.

Question two: Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the takeoff phase of the Vertical Jump Test?

Peroneus longus mean EMG activity-takeoff. As assessed by boxplot inspection, there were four outliers in the no brace condition, four outliers in the AE condition, and five outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers were found to influence the results; thus, they were removed from the analysis due to type I error concerns. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on PL mean EMG activity during the takeoff phase. The Friedman's test revealed no significant change in PL mean EMG activity during the takeoff phase, $\chi2(2) = 0.970$, $p = .616$. A one-way ANOVA for repeated measures was also conducted. There was no significant change in PL mean EMG activity during the takeoff $F(2, 64) = 2.66$, $p = .077$. Descriptive statistics are presented in Table 4.

Lateral gastrocnemius mean EMG activity-takeoff. As assessed by boxplot inspection, there were two outliers in the no brace condition. The analysis was completed with and without the outliers. These outliers were found to influence the results; thus, they were removed from the analysis due to type II error concerns. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric parametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on LG mean EMG activity during the takeoff phase. The Friedman's test revealed a significant change in LG mean EMG activity during the takeoff phase, $\chi2(2) = 9.314$, $p < .001$. A one-way ANOVA for repeated measures was also conducted. There was a similar, significant change in LG mean EMG activity during the takeoff phase, $F(2, 68) = 5.597$, $p < .001$, $\eta p^2 = .141$. Lateral gastrocnemius mean

EMG activity decreased from 73.46 +/- 34.83 %MVC when wearing no brace to 70.67 +/- 35.99 and 66.11 +/- 36.47 %MVC when wearing the AE and T1 ankle braces, respectively. Bonferroni pairwise comparisons analysis revealed a significant decrease in LG mean EMG activity when wearing the T1 ankle brace compared to no brace (-7.34 (95% CI, -13.307 to -1.376) %MVC, p = .012). There was no significant difference in LG mean EMG between no brace and AE ankle brace conditions (-2.78 (95% CI, -8.335 to 2.768) %MVC, p = .646) or AE and T1 conditions (-4.55 (95% CI, -9.751 to 0.636) %MVC, p = .102; see Figure 16).



*Significantly different from no brace condition ($p < .05$)

Figure 16. Differences in LG mean EMG activity across brace conditions.

Biceps femoris mean EMG activity-takeoff. As assessed by boxplot inspection, there was one outlier in the no brace condition, four outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on BF mean EMG activity during the takeoff phase. The Friedman's test revealed no significant change in BF mean EMG activity during the takeoff phase, $\chi2(2) = 2.450$, $p = .294$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 12.317$, $p = .002$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in BF mean EMG activity during the takeoff phase, $F(1.566, 61.088) = 1.057$, $p = .339$. Descriptive statistics are presented in Table 4.

Rectus femoris mean EMG activity-takeoff. As assessed by boxplot inspection, there was one outlier in the T1 condition. The analysis was completed with and without the outlier. The outlier did not influence the results; thus, it was included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on RF mean EMG activity during the takeoff phase. The Friedman's test revealed a significant change in RF mean EMG activity during the takeoff phase, $\chi2(2) = 21.211$, $p <.001$. Due to type I error concerns, however, a one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 6.701$, $p = .035$. Therefore, a Greenhouse-Geisser correction was applied. Unlike the Friedman's test, there was no significant change in RF mean EMG activity during the takeoff phase, $F(1.710, 63.255) = 2.845$, $p = .074$. Descriptive statistics are presented in Table 4.

Gluteus medius mean EMG activity-takeoff. As assessed by boxplot inspection, there were two outliers in the no brace condition, three outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on GM mean EMG activity during the takeoff phase. The Friedman's test revealed no significant change in GM mean EMG activity during the takeoff phase, $\chi2(2) = 2.294$, $p = .318$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 8.088$, $p = .018$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in GM mean EMG activity during the takeoff phase, $F(1.635, 53.950) = 0.377$, $p = .672$. Descriptive statistics are presented in Table 4.

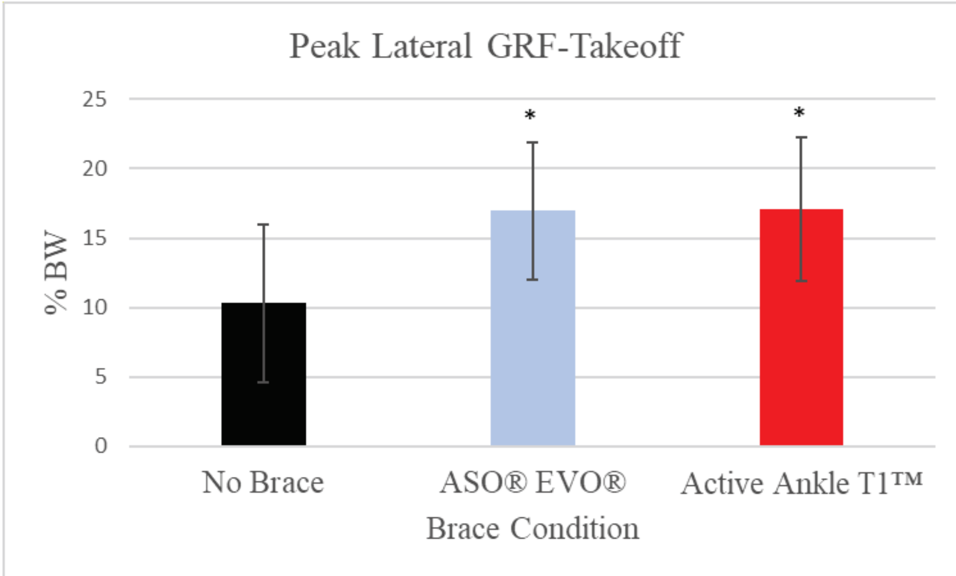
Peak vertical GRF-takeoff. As assessed by boxplot inspection, there was one outlier in the AE condition. The analysis was completed with and without the outlier. The outlier did not influence the results; thus, it was included in the analysis. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on peak vertical GRF during the takeoff phase. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 6.591$, $p = .037$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in peak vertical GRF during the takeoff phase, $F(1.725, 67.285) = 0.374$, $p = .618$. Descriptive statistics are presented in Table 4.

Peak posterior GRF-takeoff. As assessed by boxplot inspection, there were two outliers in the no brace condition, one outlier in the AE condition, and four outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed for the no brace or T1 conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). The data were normally distributed for the AE condition, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak posterior GRF during the takeoff phase. The Friedman's test revealed no significant change in peak posterior GRF during the takeoff phase, $\chi 2(2) = 4.550$, $p = .103$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 7.433$, $p = .024$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in peak posterior GRF during the takeoff phase, $F(1.698, 66.233) = 0.454$, $p = .605$. Descriptive statistics are presented in Table 4.

Peak anterior GRF-takeoff. As assessed by boxplot inspection, there were no outliers. The data were normally distributed as assessed by Shapiro-Wilk's Test ($p > .05$). A one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on peak anterior GRF during the takeoff phase. There was no significant change in peak anterior GRF during the takeoff phase, $F(2, 78) = 0.341$, $p = .712$. Descriptive statistics are presented in Table 4.

Peak lateral GRF-takeoff. As assessed by boxplot inspection, there were two outliers in the no brace condition. The analysis was completed with and without the outliers. The outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed for the no brace condition as assessed by Shapiro-Wilk's Test ($p < .05$). The data were normally distributed for the AE and T1 conditions as assessed by Shapiro-Wilk's Test ($p > .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak lateral GRF activity during the takeoff phase.

The Friedman's test revealed a significant change in peak lateral GRF during the takeoff phase, $\chi2(2) = 29.40$, $p <.001$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 59.234$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was a significant change in peak lateral GRF during the takeoff phase, $F(1.118, 43.585) = 39.80$, $p < .001$, $\eta p^2 = .505$. Peak lateral GRF increased from 10.31 +/- 5.69 %BW when wearing no brace to 16.95 +/- 4.95 and 17.07 +/- 5.18 %BW when wearing the AE and T1 ankle braces, respectively. Bonferroni pairwise comparisons analysis revealed a significant increase in peak lateral GRF when wearing the AE ankle brace compared to no brace (6.64 (95% CI, 4.066 to 9.233) %BW, $p < .001$). There was a significant increase in peak lateral GRF when wearing the T1 ankle brace compared to no brace (6.76 (95% CI, 4.131 to 9.404) %BW, $p < .001$). There was no significant difference in peak lateral GRF between AE and T1 brace conditions (0.11 (95% CI, -.60 to 0.84) %BW, $p > .05$; see Figure 17).



*Significantly different from no brace condition ($p < .05$)

Figure 17. Differences in peak lateral GRF across brace conditions.

Peak medial GRF-takeoff. As assessed by boxplot inspection, there was one outlier in the no brace, AE, and T1 condition. The analysis was completed with and without the outlier. The outlier did not influence the results; thus, it was included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak medial GRF during the takeoff phase. The Friedman's test revealed no significant change in peak medial GRF during the takeoff phase, $\chi2(2) = .950$, $p = .622$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak medial GRF during the takeoff phase, $F(2, 78) = 2.70$, $p > .05$. Descriptive statistics are presented in Table 4.

Total vertical impulse-takeoff. As assessed by boxplot inspection, there were two outliers in the no brace condition, two outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. The outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the AE condition as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total vertical impulse during the takeoff phase. The Friedman's test revealed no significant change in total vertical impulse during the takeoff phase, $\chi 2(2) = 1.40$, $p = .497$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in total vertical impulse during the takeoff phase, $F(2, 78) = 0.854$, $p > .05$. Descriptive statistics are presented in Table 4.

Landing

Reliability. Intraclass correlations for all dependent variables are presented in Table 5.

Table 5.
Intraclass correlations for Vertical Jump Test landing dependent variables across three trials.

Dependent Variable	**No Brace**	**ASO® EVO®**	**Active Ankle T1™**
EMG			
PL Mean EMG (n = 30)	.836	.662	.899
LG Mean EMG (n = 34)	.518	.634	.612
BF Mean EMG (n = 39)	.593	.666	.762
RF Mean EMG (n = 38)	.779	.542	.774
GM Mean EMG (n = 35)	.732	.636	.651
GROUND REACTION FORCE			
Peak Vertical GRF (F_z; n = 40)	.716	.774	.794
Peak Posterior GRF (F_y; n = 40)	.233	.621	.509
Peak Anterior GRF (F_y; n = 40)	.681	.663	.668
Peak Lateral GRF (F_x; n = 38)	.519	.465	.632
Peak Medial GRF (F_x; n = 49)	.618	.514	.523
Total Vertical Impulse (n = 40)	.347	.429	.651

Descriptive statistics. Descriptive statistics for all dependant variables during the landing phase of the Vertical Jump Test are presented in Table 6.

Table 6.
Electromyography and kinetic descriptive statistics for the landing phase of the Vertical Jump Test.

	No Brace			**ASO® EVO®**			**Active Ankle T1™**		
Dependant Variable	Mean	SD	n	Mean	SD	n	Mean	SD	n
EMG (%MVC)									
PL Mean EMG	56.58	19.39	30	50.06	18.76	30	52.89	21.47	30
LG Mean EMG	69.34	33.52	34	61.23	28.82	34	64.20	28.82	34
BF Mean EMG	20.16	13.61	39	19.47	10.05	39	17.93	8.53	39
RF Mean EMG	76.05	57.21	38	62.84	33.29	38	63.74	40.10	38
GM Mean EMG	31.63	21.33	35	30.12	21.02	35	28.45	18.24	35
GRF (%BW)									
Peak Vertical GRF (F_z)	205.83	55.38	40	200.91	55.91	40	208.25	59.33	40
Peak Posterior GRF (F_y)	3.01	4.34	40	3.85	6.24	40	3.58	5.17	40
Peak Anterior GRF (F_y)	19.52	7.88	40	19.46	6.94	40	19.30	8.09	40
Peak Lateral GRF (F_x)	20.61	5.39	38	19.47	5.30	38	18.02	5.41	38
Peak Medial GRF (F_x)	0.02	.27	39	0.01	.02	39	0.12	.77	39
Total Vertical Impulse (N·s)	70.43	19.8	40	63.58	19.48	40	64.09	22.14	40

Inferential statistics.

Question three: Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the landing phase of the Vertical Jump Test?

Peroneus longus mean EMG activity-landing. As assessed by boxplot inspection, there were five outliers in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers were found to influence the results; thus, they were removed from the analysis due to type I error concerns. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on PL mean EMG activity during the landing phase. There was no significant change in PL mean EMG activity during the landing phase, $F(2, 58) = 2.301$, $p = .109$. Descriptive statistics are presented in Table 6.

Lateral gastrocnemius mean EMG activity-landing. As assessed by boxplot inspection, there were two outliers in the no brace condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed in the no brace condition, as assessed by Shapiro-Wilk's Test ($p < .05$). The data were normally distributed in the AE and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on LG mean EMG activity during the landing phase. The Friedman's test revealed no significant change in LG mean EMG activity during the landing phase, $\chi 2(2) = 1.235$, $p = .539$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in LG mean EMG activity during the landing phase, $F(2.049, 66) = 2.049$, $p = .137$. Descriptive statistics are presented in Table 6.

Biceps femoris mean EMG activity-landing. As assessed by boxplot inspection, there were two outliers in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on BF mean EMG activity during the landing phase. The Friedman's test revealed no significant change in LG mean EMG activity during the landing phase, $\chi2(2) = 1.436$, $p = .488$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in BF mean EMG activity, $F(2, 76) = 1.172$, $p = .315$. Descriptive statistics are presented in Table 6.

Rectus femoris mean EMG activity-landing. As assessed by boxplot inspection, there were four outliers in the no brace condition, two outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to was conducted to compare the effect of the independent variable (brace condition) on RF mean EMG activity during the landing phase. The Friedman's test revealed no significant change in RF mean EMG activity during the landing phase, $\chi2(2) = 4.00$, $p = .135$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's test of sphericity, $\chi^2(2) = 27.023$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in RF mean EMG activity during the landing phase, $F(1.309, 48.431) = 2.466$, $p = .114$. Descriptive statistics are presented in Table 6.

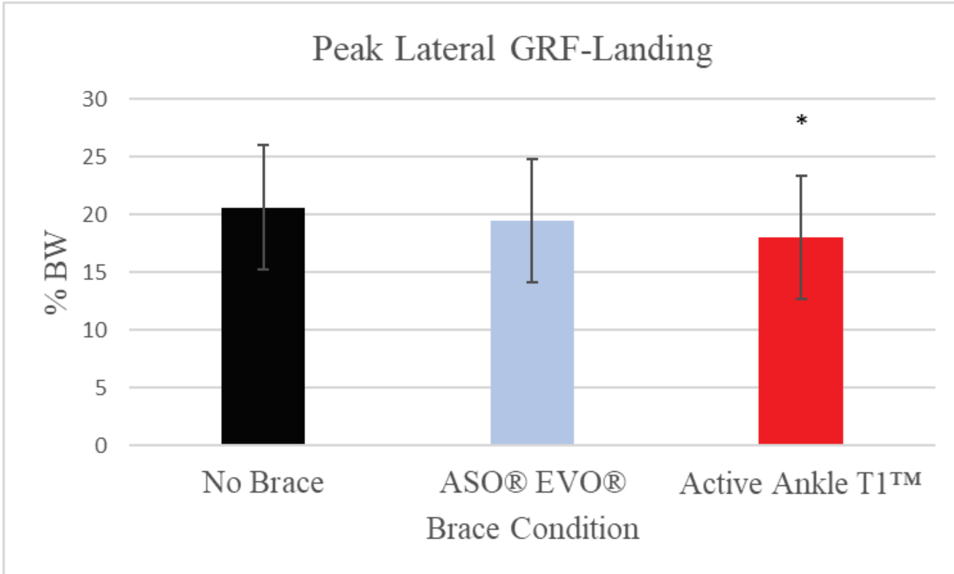
*Gluteus medius mean EMG activity-landing***.** As assessed by boxplot inspection, there were two outliers in the no brace condition and three outliers in the AE condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The were not normally distributed in the no brace or AE conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). The data were normally distributed in the T1 condition, as assessed by Shapiro-Wilk's Test. As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on GM mean EMG activity during the landing phase. The Friedman's test revealed no significant change in GM mean EMG activity during the landing phase, $\chi2(2) = 1.771$, $p = .412$. As with the Friedman's test, there was no significant change in GM mean EMG activity, $F(2, 68) = 0.800$, $p = .800$. Descriptive statistics are presented in Table 6.

Peak Vertical GRF-landing. As assessed by boxplot inspection, there was one outlier in the AE condition. The analysis was completed with and without the outlier .This outlier did not influence the results; thus, it was included in the analysis. The data were normally distributed in the no brace condition, as assessed by Shapiro-Wilk's Test ($p > .05$). This outlier did not influence the results; thus, it was included in the analysis. The data were not normally distributed in the AE and T1 condition, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak vertical GRF during the landing phase. The Friedman's test revealed no significant change in peak vertical GRF during the landing phase, $\chi2(2) = 1.850$, $p = .397$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak vertical GRF, $F(2, 78) = 0.561$, $p = .573$. Descriptive statistics are presented in Table 6.

Peak posterior GRF-landing. As assessed by boxplot inspection, there were two outliers in the no brace condition and five outliers in the AE condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak posterior GRF during the landing phase. The Friedman's test revealed no significant change in peak posterior GRF during the landing phase, $\chi2(2) = 0.450$, $p = .799$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's test of sphericity, $\chi^2(2) = 14.618$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in peak posterior GRF during the landing phase, $F(1.516, 59.120) = 0.623$, $p = .497$. Descriptive statistics are presented in Table 6.

Peak anterior GRF-landing. As assessed by boxplot inspection, there was one outlier in the no brace condition and one outlier in the AE condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the AE condition, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak anterior GRF during the landing phase. The Friedman's test revealed no significant change in peak anterior GRF during the landing phase, $\chi2(2) = 0.450$, $p = .799$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak anterior GRF during the landing phase, $F(2, 78) = 0.040$, $p = .961$. Descriptive statistics are presented in Table 6.

Peak lateral GRF-landing. As assessed by boxplot inspection, there was one outlier in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers were found to influence the results; thus, they were removed from the analysis due to type II error concerns. The data were normally distributed as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on peak lateral GRF during the landing phase. There was a significant change in peak lateral GRF during the landing phase, $F(2, 74) = 5.746$, $p = .005$, partial $\eta^2 = .134$. Peak lateral GRF decreased from 20.61 +/- 5.39 %BW when wearing no brace to 19.47 +/- 5.30 and 18.02 +/- 5.41 %BW when wearing the AE and T1 ankle braces, respectively. Bonferroni pairwise comparisons analysis revealed a significant decrease in peak lateral GRF when wearing the T1 ankle brace compared to no brace (-2.59 (95% CI, -4.432 to -0.764) %BW, $p = .003$). There was no significant difference in peak lateral GRF between no brace and AE conditions (-1.40 (95% CI, -.3.20 to 0.304) %BW, $p = .135$), or AE and T1 conditions (-1.14 (95% CI, -3.314 to 1.108) %BW, $p > .05$; see Figure 18).



*Significantly different from no brace condition ($p < .05$)

Figure 18. Differences in peak lateral GRF across brace conditions.

Peak medial GRF-landing. As assessed by boxplot inspection, there were six outliers in the no brace condition, four outliers in the AE condition, and seven outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak medial GRF during the landing phase. The Friedman's test revealed no significant change in peak medial GRF during the landing phase, $\chi 2(2) = 0.555$, $p = .758$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi^2(2) = 41.334$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in peak medial GRF during the landing phase, $F(1.196, 45.433) = 0.931$, $p = .356$. Descriptive statistics are presented in Table 6.

Total vertical impulse-landing. As assessed by boxplot inspection, there were two outliers in the AE condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on total vertical impulse during the landing phase. There was no significant change in total vertical impulse during the landing phase, $F(2, 78) = 3.265$, $p = .043$. Descriptive statistics are presented in Table 6.

Chapter Four: Lateral Cutting Task Results

Missing Data

Due to equipment malfunction (electrode failure, loss of electrode adherence, force platform failure), usable data for each variable varies from n = 26 to n = 34. In cases where only two out of three trials were considered usable due to equipment malfunction, the mean of the two trials was used in place of a third trial when calculating the average of the three trials.

Deceleration

Reliability. Intraclass correlations for all dependent variables are presented in Table 7.

Table 7.
Intraclass correlations for cutting task performance and deceleration dependent variables across three trials.

Dependent Variable	No Brace	ASO® EVO®	Active Ankle T1™
PERFORMANCE			
Lateral Cutting Task Time (n = 39)	.806	.967	.970
EMG			
PL Mean EMG (n = 33)	.725	.650	.762
LG Mean EMG (n = 30)	.413	.736	.508
BF Mean EMG (n = 30)	.735	.584	.650
RF Mean EMG (n = 29)	.431	.670	.508
GM Mean EMG (n = 26)	.553	.786	.628
GROUND REACTION FORCE			
Peak Vertical GRF (F_z; n = 34)	.796	.833	.798
Peak Posterior GRF (F_y; n = 34)	.202	.643	.324
Peak Anterior GRF (F_y; n = 34)	.890	.254	.824
Peak Lateral GRF (F_x; n = 34)	.567	.697	.675
Peak Medial GRF (F_x; n = 34)	.276	.482	.472
Total Vertical Impulse (n = 34)	.769	.640	.772
Total Anteroposterior Impulse (n = 34)	.341	.413	.436
Total Mediolateral Impulse (n = 34)	.339	.445	.300

Descriptive statistics. Descriptive statistics for all dependant variables during the deceleration phase of the cutting task are presented in Table 8.

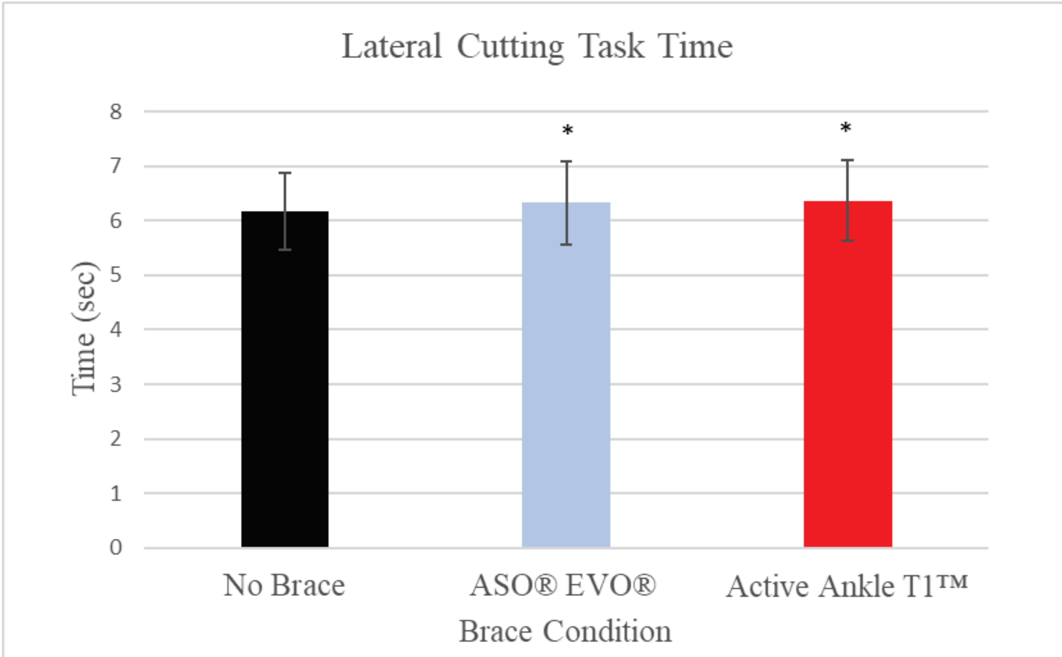
Table 8.
Electromyography and kinetic descriptive statistics for the deceleration phase of the cutting task.

	No Brace			ASO® EVO®			Active Ankle T1™		
Dependant Variable	Mean	SD	n	Mean	SD	n	Mean	SD	n
EMG (%MVC)									
PL Mean EMG	83.38	41.64	33	81.28	39.69	33	81.12	48.42	33
LG Mean EMG	52.77	44.78	30	54.21	34.53	30	56.09	29.59	30
BF Mean EMG	42.96	15.53	30	40.27	15.95	30	43.74	20.25	30
RF Mean EMG	100.98	78.33	29	86.72	64.15	29	99.45	81.96	29
GM Mean EMG	78.99	49.98	26	69.63	40.69	26	64.44	26.83	26
GRF (%BW)									
Peak Vertical GRF (F_z)	151.61	64.31	34	150.24	66.55	34	149.61	74.45	34
Peak Posterior GRF (F_y)	1.28	2.35	34	1.31	3.08	34	1.03	2.14	34
Peak Anterior GRF (F_y)	80.23	27.22	34	78.54	27.85	34	79.24	25.25	34
Peak Lateral GRF (F_x)	28.05	8.20	34	27.03	10.81	34	27.67	9.64	34
Peak Medial GRF (F_x)	1.40	1.62	34	1.50	1.62	34	1.41	1.59	34
Total Vertical Impulse (N·s)	94.06	38.64	34	89.92	38.15	34	93.14	36.13	34
Total Anteroposterior Impulse (N·s)	40.21	12.89	34	36.01	11.37	34	39.76	13.66	34
Total Mediolateral Impulse (N·s)	7.72	3.45	34	6.17	3.37	34	7.49	3.30	34

Inferential statistics.

Question four: Is there a difference amongst ankle brace conditions in time to complete the cutting task?

As assessed by boxplot inspection, there was one outlier in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on lateral cutting task time. There was a significant change in time to complete the lateral cutting task, $F(2, 76) = 17.242$, $p < .001$, $\eta p^2 = .312$. Mean time to complete the cutting task increased from 6.17 +/- .71 sec when wearing no brace to 6.33 +/- .76 and 6.37 +/- 0.75 sec when wearing the AE and T1 ankle braces, respectively. Bonferroni pairwise comparisons analysis revealed a significant increase in time to complete the cutting task when wearing the AE ankle brace compared to no brace (0.16 (95% CI, .062 to .265) sec, $p < .001$). Additionally, there was a significant increase in time to complete the cutting task when wearing the T1 ankle brace compared to no brace (0.2(95% CI, .113 to .286) sec, $p < .001$). There was no significant difference in time to complete the cutting task between AE and T1 ankle brace conditions (0.03 (95% CI, -.048 .to .120) sec, $p > .05$; see Figure 19).



*Significantly different from no brace condition ($p < .05$)

Figure 19. Differences in lateral cutting task time across brace conditions.

Question five: Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the deceleration phase of the cutting task?

Peroneus longus mean EMG activity-deceleration. As assessed by boxplot inspection, there were three outliers in the no brace condition, two outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on PL mean EMG activity during the deceleration phase. The Friedman's test revealed no significant change in PL mean EMG activity during the deceleration phase, $\chi2(2) = 0.182$, $p = .913$. A one-way ANOVA for repeated measures was also conducted. As with the

Friedman's test, there was no significant change in PL mean EMG activity during the deceleration phase, $F(2, 64) = 0.227$, $p = .798$. Descriptive statistics are presented in Table 8.

Lateral gastrocnemius mean EMG activity-deceleration. As assessed by boxplot inspection, there were two outliers in the no brace condition and one outlier in the AE condition. The outlier in the AE condition was removed from the analysis due to measurement error concerns. The analysis was completed with and without the remaining outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on LG mean EMG activity during the deceleration phase. The Friedman's test revealed no significant change in LG mean EMG activity during the deceleration phase, $\chi 2(2) = 4.867$, $p = .088$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in LG mean EMG activity during the deceleration phase, $F(2, 58) = 0.182$, $p = .834$. Descriptive statistics are presented in Table 8.

Biceps femoris mean EMG activity-deceleration. As assessed by boxplot inspection, there was one outlier in the no brace condition and one outlier in the AE condition. The data were normally distributed in the T1 condition, as assessed by Shapiro-Wilk's Test ($p > .05$). The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed in the no brace and AE conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on BF mean EMG activity during the deceleration phase. The Friedman's test revealed no significant change in BF mean EMG activity during the deceleration phase, $\chi2(2) = 2.40$, $p = .301$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in BF mean EMG activity during the deceleration phase, $F(2, 58) = 0.690$, $p = .505$. Descriptive statistics are presented in Table 8.

Rectus femoris mean EMG activity-deceleration. As assessed by boxplot inspection, there were four outliers in the no brace condition, one outlier in the AE condition, and four outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on RF mean EMG activity during the deceleration phase. The Friedman's test revealed no significant change in RF mean EMG activity during the deceleration phase, $\chi2(2) = 0.276$, $p = .871$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 13.645$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in RF mean EMG activity during the deceleration phase, $F(1.271, 35.95) = 1.449$, $p =.243$. Descriptive statistics are presented in Table 8.

Gluteus medius mean EMG activity-deceleration. As assessed by boxplot inspection, there were two outliers in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the T1 condition, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the no brace and AE conditions, as assessed by Shapiro-Wilk's Test ($p <$

.05 As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on GM mean EMG activity during the deceleration phase. The Friedman's test revealed no significant change in GM mean EMG activity during the deceleration phase, $\chi2(2) = 3.308$, $p = .191$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in GM mean EMG activity during the deceleration phase, $F(2, 50) = 2.281$, $p =.113$. Descriptive statistics are presented in Table 8.

Peak vertical GRF-deceleration. As assessed by boxplot inspection, there were four outliers in the no brace condition, three outliers in the AE condition, and four outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak vertical GRF during the deceleration phase. The Friedman's test revealed no significant change in peak vertical GRF during the deceleration phase, $\chi2(2) = 0.765$, $p = .682$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak vertical GRF during the deceleration phase, $F(2, 66) = 0.105$, $p = .901$. Descriptive statistics are presented in Table 8.

*Peak posterior GRF-deceleration***.** As assessed by boxplot inspection, there were three outliers in the no brace condition, three outliers in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$ As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak posterior GRF during the deceleration phase. The Friedman's test revealed no significant change in peak posterior GRF during the deceleration phase, $\chi2(2) = 0.235$, $p = .889$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 9.436$, $p = .009$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in peak posterior GRF during the deceleration phase, $F(1.603, 52.574) = 0.331$, $p = .670$. Descriptive statistics are presented in Table 8.

Peak anterior GRF-deceleration. As assessed by boxplot inspection, there was one outlier in the no brace condition. The analysis was completed with and without the outlier. This outlier did not influence the results; thus, is was included in the analysis. The data were normally distributed in the AE condition, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak anterior GRF during the deceleration phase. The Friedman's test revealed no significant change in peak anterior GRF during the deceleration phase, $\chi2(2) = 2.882$, $p = .237$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak anterior GRF during the deceleration phase, $F(2, 66) = 0.385$, $p = .682$. Descriptive statistics are presented in Table 8.

Peak lateral GRF-deceleration. As assessed by boxplot inspection, there were four outliers in the no brace condition, two outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test (*p* < .05 As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak lateral GRF during the deceleration phase. The Friedman's test revealed no significant change in peak lateral GRF during the deceleration phase, $\chi2(2) = 0.412$, $p = .814$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak lateral GRF during the deceleration phase, $F(2, 66) = 0.354$, $p = .704$. Descriptive statistics are presented in Table 8.

Peak medial GRF-deceleration. As assessed by boxplot inspection, there were two outliers in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test (*p* < .05 As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak medial GRF during the deceleration phase. The Friedman's test revealed no significant change in peak medial GRF during the deceleration phase, $\chi2(2) = 1.647$, $p = .439$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak medial GRF during the deceleration phase, $F(2, 66) = .069$, $p = .933$. Descriptive statistics are presented in Table 8.

Total vertical impulse-deceleration. As assessed by boxplot inspection, there were three outliers in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$).). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total vertical impulse during the deceleration phase. The Friedman's test revealed no significant change in total vertical impulse during the deceleration phase, $\chi2(2) = 2.176$, $p = .337$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in total vertical impulse during the deceleration phase, $F(2, 66) = 1.285$, $p = .283$. Descriptive statistics are presented in Table 8.

Total anteroposterior impulse-deceleration. As assessed by boxplot inspection, there was one outlier in the AE condition and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the AE condition, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total anteroposterior impulse during the deceleration phase. The Friedman's test revealed no significant change in total anteroposterior impulse during the deceleration phase, $\chi2(2) = 4.647$, $p = .098$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 7.491$, $p = .024$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in total anteroposterior impulse during the deceleration phase, $F(1.655, 54.604) = 4.161$, $p = .27$. Descriptive statistics are presented in Table 8.

Total mediolateral impulse-deceleration. As assessed by boxplot inspection, there was one outlier in the no brace condition, one outlier in the AE condition, and five outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the AE and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the no brace condition, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total mediolateral impulse during the deceleration phase. The Friedman's test revealed no significant change in total mediolateral impulse during the deceleration phase, $\chi2(2) = 5.706$, $p = .058$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in total mediolateral impulse during the deceleration phase, $F(2, 66) = 3.802$, $p = .027$. Descriptive statistics are presented in Table 8.

Propulsive

Reliability. Intraclass correlations for all dependent variables are presented in Table 9.

Table 9.
Intraclass correlations for deceleration dependent variables across three trials.

Dependent Variable	No Brace	ASO® EVO®	Active Ankle T1™
EMG			
PL Mean EMG (n = 33)	.876	.856	.876
LG Mean EMG (n = 30	.842	.847	.439
BF Mean EMG (n = 30)	.853	.838	.760
RF Mean EMG (n = 29)	.754	.588	.746
GM Mean EMG (n = 26)	.831	.800	.654
GROUND REACTION FORCE			
Peak Vertical GRF (F_z; n = 34)	.980	.986	.978
Peak Posterior GRF (F_y; n = 34)	.610	.576	.344
Peak Anterior GRF(F_y; n = 34))	.890	.592	.825
Peak Lateral GRF (F_x; n = 34)	.780	.847	.780
Peak Medial GRF (F_x; n = 34)	.342	.234	.472
Total Vertical Impulse (n = 34)	.772	.862	.867
Total Anteroposterior Impulse (n = 34)	.731	.839	.714
Total Mediolateral Impulse (n = 34)	.485	.825	.733

Descriptive statistics. Descriptive statistics for all dependant variables during the propulsive phase of the cutting task are presented in Table 10.

Table 10.
Electromyography and kinetic descriptive statistics for the propulsive phase of the cutting task.

Dependant Variable	No Brace Mean	SD	n	ASO® EVO® Mean	SD	n	Active Ankle T1™ Mean	SD	n
EMG (%MVC)									
PL Mean EMG	68.59	28.40	33	68.92	30.11	33	70.59	36.48	33
LG Mean EMG	53.07	23.46	30	53.99	24.62	30	58.17	36.04	30
BF Mean EMG	37.54	16.64	30	38.81	19.69	30	39.73	19.48	30
RF Mean EMG	68.46	32.63	29	63.71	27.18	29	67.77	33.79	29
GM Mean EMG	59.44	28.82	26	62.55	29.03	26	58.17	25.75	26
GRF (%BW)									
Peak Vertical GRF (F_z)	135.16	48.85	34	135.62	44.64	34	137.13	47.41	34
Peak Posterior GRF (F_y)	2.10	1.94	34	2.26	1.91	34	1.82	1.75	34
Peak Anterior GRF (F_y)	80.27	27.23	34	79.43	27.75	34	79.24	25.25	34
Peak Lateral GRF (F_x)	59.25	11.39	34	60.42	10.35	34	60.39	9.65	34
Peak Medial GRF (F_x)	1.32	2.06	34	2.08	1.132	34	2.085	1.08	34
Total Vertical Impulse (N·s)	297.89	114.02	34	318.78	156.50	34	315.84	131.39	34
Total Anteroposterior Impulse (N·s)	114.17	34.03	34	118.96	44.08	34	114.96	37.60	34
Total Mediolateral Impulse (N·s)	77.82	20.88	34	84.02	19.19	34	81.33	18.25	34

Inferential statistics.

Question six: Is there a difference amongst ankle brace conditions in measures of lower extremity EMG activity and kinetics during the propulsive phase of the cutting task?

Peroneus longus mean EMG activity-propulsive. As assessed by boxplot inspection, there were two outliers in the no brace condition, four outliers in the AE condition, and three outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on PL mean EMG activity during the propulsive phase. The Friedman's test revealed no significant change in PL mean EMG activity during the deceleration phase, $\chi2(2) = 1.267$, $p = .531$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 7.787$, $p = .020$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in PL mean EMG activity during the propulsive phase, $F(1.637, 52.368) = 0.457$, $p = .597$. Descriptive statistics are presented in Table 10.

Lateral gastrocnemius mean EMG activity-propulsive. As assessed by boxplot inspection, there was one outlier in the T1 condition. The analysis was completed with and without the outlier. The outlier did not influence the results; thus, it was included in the analysis. The data were normally distributed in the no brace and AE conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the T1 condition, as assessed by Shapiro-Wilk's Test ($p < .05$ As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on LG mean EMG activity during the propulsive phase. The Friedman's test revealed no significant change in LG mean EMG activity during the propulsive phase, $\chi2(2) = 1.40$, $p = .497$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 27.365$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in LG mean

EMG activity during the propulsive phase, $F(1.232, 35.724) = 1.088$, $p = .318$. Descriptive statistics are presented in Table 10.

Biceps femoris mean EMG activity-propulsive. As assessed by boxplot inspection, there was one outlier in the no brace condition, two outliers in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on BF mean EMG activity during the propulsive phase. The Friedman's test revealed no significant change in LG mean EMG activity during the propulsive phase, $\chi2(2) = 0.467$, $p = .792$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in BF mean EMG activity during the propulsive phase, $F(2, 58) = 0.397$, $p = .674$. Descriptive statistics are presented in Table 10.

Rectus femoris mean EMG activity-propulsive. As assessed by boxplot inspection, there was one outlier in the no brace condition and one outlier in the AE condition. Upon inspection of the raw data, these outliers were removed due to measurement error concerns. The data were normally distributed in the AE condition, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on RF mean EMG activity during the propulsive phase. The Friedman's test revealed no significant change in RF mean EMG activity during the propulsive phase, $\chi2(2) = 0.483$, $p = .786$. As with the Friedman's test, there was no significant change in RF mean EMG activity during the propulsive phase, $F(2, 56) = 1.027$, $p = .365$. Descriptive statistics are presented in Table 10.

Gluteus medius mean EMG activity-propulsive. As assessed by boxplot inspection, there was one outlier in the no brace condition, one outlier in the AE condition, and one outlier in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on GM mean EMG activity during the propulsive phase. The Friedman's test revealed no significant change in GM mean EMG activity during the propulsive phase, $\chi2(2) = 2.769$, $p = .250$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in GM mean EMG activity during the propulsive phase, $F(2, 50) = 1.158$, $p = .322$. Descriptive statistics are presented in Table 10.

*Peak vertical GRF-propulsive***.** As assessed by boxplot inspection, there were three outliers in the no brace condition, four outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak vertical GRF during the propulsive phase. The Friedman's test revealed no significant change in peak vertical GRF during the propulsive phase, $\chi2(2) = 0.706$, $p = .703$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in peak vertical GRF during the propulsive phase, $F(2, 66) = 0.821$, $p = .444$. Descriptive statistics are presented in Table 10.

Peak posterior GRF-propulsive. As assessed by boxplot inspection, there were two outliers in the no brace condition and two outliers in the T1 condition. . The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on peak posterior GRF during the propulsive phase. There was no significant change in peak posterior GRF during the propulsive phase, $F(2, 66) = 1.896$, $p = .158$. Descriptive statistics are presented in Table 10.

*Peak anterior GRF-propulsive***.** As assessed by boxplot inspection, there was one outlier in the no brace condition. The analysis was completed with and without the outlier. This outlier did not influence the results; thus, it was included in the analysis. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on peak anterior GRF during the propulsive phase. There was no significant change in peak anterior GRF during the propulsive phase, $F(2, 66) = .180$ $p = .835$. Descriptive statistics are presented in Table 10.

Peak lateral GRF-propulsive. As assessed by boxplot inspection, there were two outliers in the no brace condition, three outliers in the AE condition, and three outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed, as assessed by Shapiro-Wilk's Test ($p > .05$). As such, a one-way ANOVA for repeated measures was conducted to compare the effect of the independent variable (brace condition) on peak lateral GRF during the propulsive phase. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 7.962$, $p = .019$. Therefore, a Greenhouse-Geisser correction was applied. There was no significant change in peak lateral GRF during the propulsive phase, $F(1.639, 54.087) = .869$, $p = .406$. Descriptive statistics are presented in Table 10.

Peak medial GRF-propulsive. As assessed by boxplot inspection, there were four outliers in the no brace condition, three outliers in the AE condition, and three outliers in the T1 condition. . The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the T1 condition, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the no brace and AE conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on peak medial GRF during the propulsive phase. The Friedman's test revealed no significant change in peak medial GRF during the propulsive phase, $\chi2(2) = 3.059$, $p = .217$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 35.299$, $p < .001$. Therefore, a Greenhouse-Geisser correction was applied As with the Friedman's test, there was no significant change in peak medial GRF during the propulsive phase, $F(1.199, 39.565) = 2.898$, $p = .09$. Descriptive statistics are presented in Table 10.

Total vertical impulse-propulsive. As assessed by boxplot inspection, there were three outliers in the no brace condition, two outliers in the AE condition, and two outliers in the T1 condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were not normally distributed, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total vertical impulse during the propulsive phase. The Friedman's test revealed no significant change in total vertical impulse during the propulsive phase, $\chi2(2) = 2.176$, $p = .337$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in total vertical impulse during the propulsive phase, $F(2, 66) = 2.615$, $p = .081$. Descriptive statistics are presented in Table 10.

Total anteroposterior impulse-propulsive. As assessed by boxplot inspection, there was one outlier in the T1 condition. The analysis was completed with and without the outlier. This outlier did not influence the results; thus, it was included in the analysis. The data were normally distributed in the no brace and T1 condition, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the AE and T1 conditions, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total anteroposterior impulse during the propulsive phase. The Friedman's test revealed no significant change in total anteroposterior impulse during the propulsive phase, $\chi2(2) = 1.588$, $p = .452$. A one-way ANOVA for repeated measures was also conducted. As with the Friedman's test, there was no significant change in total anteroposterior impulse during the propulsive phase, $F(2, 66) = 1.023$, $p = .365$. Descriptive statistics are presented in Table 10.

Total mediolateral impulse-propulsive. As assessed by boxplot inspection, there was one outlier in the no brace condition and one outlier in the AE condition. The analysis was completed with and without the outliers. These outliers did not influence the results; thus, they were included in the analysis. The data were normally distributed in the no brace and T1 conditions, as assessed by Shapiro-Wilk's Test ($p > .05$). The data were not normally distributed in the AE condition, as assessed by Shapiro-Wilk's Test ($p < .05$). As such, nonparametric and parametric testing was conducted to compare the effect of the independent variable (brace condition) on total mediolateral impulse during the propulsive phase. The Friedman's test revealed no significant change in total mediolateral impulse during the propulsive phase, $\chi2(2) = 1.235$, $p = .539$. A one-way ANOVA for repeated measures was also conducted. The assumption of sphericity was not met, as assessed by Mauchly's Test of Sphericity, $\chi2(2) = 11.687$, $p = .003$. Therefore, a Greenhouse-Geisser correction was applied. As with the Friedman's test, there was no significant change in total mediolateral impulse during the propulsive phase, $F(1.531, 50.537) = 2.580$, $p = .099$. Descriptive statistics are presented in Table 10.

Chapter Five: Discussion

The purpose of the current study was to determine if wearing semi-rigid or softshell ankle braces affects vertical jump height and time to complete a cutting task, as well as kinetics and lower extremity EMG activity during these measures in healthy, active individuals with jumping and running sport experience. Although previous literature examining the effects of ankle bracing on athletic performance is conflicting, the results of the current study lend support to the notion that ankle braces significantly decreases athletic performance. Additionally, while the body of research examining the effects of ankle braces on EMG and kinetic variables during athletic performance is limited, the results of the current study suggest that ankle bracing can significantly affect these variables during vertical jumping. The following section will discuss the results and implications of the current study in greater detail.

Takeoff and Flight Phase

Compared to wearing no ankle braces, there was a significant reduction of 2.1 cm and 2.13 cm in vertical jump height when wearing AE and T1 ankle braces, respectively. The performance results of the Vertical Jump Test are in line with recent research by Henderson et al. (2016), Smith et al. (2016), and Parsley et al. (2013). Henderson et al. did not observe a significant change in vertical jump height when wearing AE ankle braces, although there was a mean decrease of 1.47 cm when wearing the AE ankle braces. They did, however, observe a significant decrease in vertical jump height of 2.35 cm when wearing the T1 ankle brace, compared to the no brace condition. Smith et al. recorded a similar reduction in vertical jump height of 2.33 cm, albeit when wearing McDavid 195T softshell ankle braces. Furthermore, Parsley et al. observed a significant decrease in vertical jump height of 1.4 cm when wearing both the AE ankle brace and Aircast Airsport™ semi-rigid ankle brace.

As softshell ankle braces function to restrict motion in both the frontal and sagittal planes (Gudibanda & Wang, 2005), the reduction in vertical jump height observed in the current study when wearing softshell ankle braces, as well as others, is not unexpected. Although the musculature producing movement at the ankle joint is not a large generator of power relative to the musculature of the knee and hip (Prilutsky & Zatsiorsky, 1994), the ankle joint represents the last major joint in a proximal to distal model of muscular activation and joint reversals during a maximum vertical jump (Pandy & Zajac, 1991). As a biarticulated muscle performing both knee flexion and ankle plantarflexion (Tortora & Nielson, 2010), the gastrocnemius muscle plays a role in transferring mechanical energy from the hip and knee to the ankle during a vertical jump (Prilutsky & Zatsiorsky, 1994). Considering ankle braces have been demonstrated to reduce ankle ROM during dynamic jumping tasks (Cordova et al., 2010; DiStefano et al., 2008; Hodgson et al., 2005; Simpson et al., 2003; Smith et al., 2016), theoretically this would reduce the elongation and stretch of the gastrocnemius muscle during the eccentric portion of the vertical jump, thereby reducing the length from which a concentric contraction of the gastrocnemius muscle would be initiated. Based on the length-tension relationship, this would reduce the amount of force that could be generated and transferred, as contribution of the parallel and series elastic components to tension development is decreased via reduced elongation of the gastrocnemius muscle (Hamill, Knutzen, & Derrick, 2015). In turn, vertical jump height would be reduced.

With respect to the decrease in LG mean EMG activity observed when wearing the T1 ankle braces but not the AE ankle braces, two explanations are possible. Previous literature has suggested that plantar-flexor EMG activity is reduced when wearing an ankle brace, due to an ankle brace restricting ankle ROM (Smith et al., 2016). Therefore, it is possible that the T1 ankle brace restricts plantarflexion more than the AE ankle brace, resulting in decreased LG activation. Alternatively, although not providing more plantarflexion restriction than the AE ankle brace, the design of the T1 ankle brace, specifically the heel pad and hinge, may influence the EMG activity of the LG muscle by altering sensory feedback. As studies have shown that softshell ankle braces, such as the AE ankle brace, generally restrict motion in the sagittal plane more than semi-rigid ankle braces during dynamic tasks (Cordova et al., 2010; Tamura et al., 2017), the latter may be more likely; support for this rationale can be found in research examining the effects of cutaneous and tactile stimulation on musculature of the of the foot during gait.

Many receptors, including Meissner's corpuscles, Pacinian corpuscles, Merkel disk receptors, and Ruffini endings are located on the plantar aspect of the foot, therefore, the heel and sole represent a highly sensitive area to cutaneous and tactile stimulation (Zehr et al., 2014). As such, the cutaneous receptors of the foot will detect changes in contact pressure (Magnusson, Enbom, Johansson, & Pyykkö, 1990). In combination with feedback from mechanoreceptors and nociceptors, this information serves to modulate reflexes that contribute to and are involved in locomotion (Kennedy & Inglis, 2002). Previous research has noted that decreased tactile stimulation of the heel facilitates medial gastrocnemius muscle activity during gait (Nurse & Nigg, 2001). On the other hand, increased tactile stimulation during the stance phase of gait may result in corrective postural changes, including facilitation of the TA muscle (Zehr et al., 2014), which is responsible for dorsiflexion and inversion of the ankle (Tortora & Nielson, 2010). As such, it is logical to suggest that the heel pad of the T1 ankle brace would alter cutaneous feedback of the heel, and, if the TA muscle was facilitated, this would likely result in inhibition of the antagonist muscle, the gastrocnemius (Tortora & Nielson, 2010). As TA EMG was not measured in the current study, it is difficult to ascertain if this facilitation/inhibition interaction occurred; however, as LG mean EMG activity did decrease in the T1 ankle brace condition during takeoff, relative to no brace, this rationale may provide some explanation as to why this occurred.

While no significant change in LG mean EMG activity was observed in the current study when wearing the AE softshell ankle braces during the takeoff phase of the Vertical Jump Test, there was a slight decrease of 2.78 %MVC relative to the no brace condition. Additionally, there was a tendency towards a decrease in PL mean EMG activity of 3.8 %MVC when wearing AE ankle braces. Since PL is a synergistic muscle that assists with plantarflexion (Tortora & Nielson, 2010), this would suggest and support the notion that the length of the muscle, and, therefore, EMG activity, is being reduced as a result of softshell ankle braces restricting motion in the sagittal plane (Smith et al., 2016). Smith et al. (2016) noted a similar trend in the gastrocnemius muscle EMG activity when wearing softshell ankle braces as the current study. They also noted a significant reduction in soleus muscle EMG activity, another synergistic muscle to plantarflexion, during a vertical jump test when wearing softshell ankle braces. They postulated that these reductions in plantar-flexor muscle activity and decreased ankle ROM contributed to the reduction in vertical jump height when wearing the softshell ankle braces. As Pandy and Zajac (1991) noted that the gastrocnemius muscle can be responsible for up to 25% of vertical jump height, the overall reduction in plantar-flexor muscle activity concurrently with vertical jump height would suggest a relationship. The exact mechanism behind the reduction in vertical jump height, however, remains unclear when examining the kinetic data of the current study. Although the reduction of plantar-flexor muscle activity and ROM should theoretically reduce the amount of force transferred and generated at the ankle, the results of the current study do not support this rationale. When wearing the AE softshell ankle braces, peak vertical GRF was not significantly affected during the takeoff phase of the Vertical Jump Test, nor was total vertical impulse. Given the small decrease in plantar-flexor muscle activity observed in the current study and similar vertical GRF across conditions, it is unlikely that this reduction contributed to a reduction in overall vertical jump height.

The likelihood of a small reduction in plantarflexion ROM contributing to a reduction in vertical jump height is further decreased when examining the effects of a semi-rigid ankle braces on vertical jump height. Although studies are conflicting with respect to semi-rigid ankle braces ability to restrict sagittal plane motion, semi-rigid ankle braces are designed to allow for unrestricted movement in the sagittal plane (i.e., plantarflexion and dorsiflexion; Mackean et al., 1995) relative to the AE ankle brace. As predictive models of vertical jump performance predominantly involve sagittal plane kinematics and kinetics (Aragón-Vargas & Gross, 1997), theoretically, semi-rigid ankle braces should not affect kinematic factors that would contribute to a reduction in vertical jump height. In the current study, however, the T1 ankle braces reduced vertical jump height more than the AE ankle braces relative to the no brace condition, although theoretically allowing for greater plantarflexion and dorsiflexion ROM (Mackean et al., 1995). This reduction in vertical jump height was also accompanied by a significant reduction in LG mean EMG activity of 7.34% of MVC, which was not present in the AE condition. Again, peak vertical and total vertical impulse were not affected during the takeoff phase of the Vertical Jump Test, indicating that the decrease in vertical jump height was not the result of decreased force production at the ankle.

Although no changes were observed in measures of vertical GRF, there was a significant increase in peak lateral GRF during the takeoff phase of the Vertical Jump Test when wearing both the AE and T1 ankle braces, compared to the no brace condition. This increase amounted to

6.64 %BW when wearing the AE softshell ankle braces and 6.76 %BW when wearing the T1 semi-rigid ankle braces. These results contradict Castro et al. (2017), who found no differences between Horse® softshell ankle braces and no brace conditions in peak medial or lateral GRF during a vertical jump takeoff. Given that the Horse® ankle brace is very similar in design to the AE ankle brace, the differing results may likely be due to methodological designs of the studies. Castro et al. utilized a submaximal, countermovement jump, where as the current study utilized a standardized maximum Vertical Jump Test involving a pause while in 45° of knee flexion. As such, the contribution of the stretch shortening cycle was reduced, while an isometric component was introduced to the movement. Therefore, any potential biomechanical changes as a result of wearing an ankle brace may have been accentuated while the participant isometrically held this position, resulting in altered GRFs during takeoff. As only peak lateral GRF was affected, however, this would likely suggest biomechanical changes in the frontal plane and not the sagittal plane. In the absence of frontal plane kinematic data from any study for the hip and knee during a vertical jump takeoff, the known effects of ankle braces on frontal plane ankle kinematics, in combination with the EMG data from the current study, should be considered when trying to explain the increase in peak vertical GRF.

During normal gait, both pronation and supination of the subtalar joint occur. At immediate heel strike, the subtalar joint will be supinated; however, as the foot moves to full ground contact, the subtalar joint will pronate (Tiberio, 1987). This is followed once again by supination through the midstance phase to toe off (Tiberio, 1987). In order for the subtalar joint to pronate during ground contact, the talus will move medially while the calcaneus everts (Root, Orien, & Weed, 1977). To allow this motion of the talus within the subtalar joint, the entire lower extremity must internally rotate during pronation, and externally rotate during supination

(Inman, 1976; Kirby, 2000). By restricting supination ROM at the ankle, ankle braces may prevent this reversal of internal rotation. During a cutting task, Klem et al. (2016) noted that both AE ankle braces and Active Ankle T2™ semi-rigid ankle braces significantly increased internal rotation at the knee, while creating a clinically relevant increase in knee abduction angle. Although frontal and sagittal plane kinematic data were not collected in the current study, it is possible to infer that these effects may be present at the knee when wearing AE ankle braces during the Vertical Jump Test. While not significantly different, there was a 7.16% of MVC decrease in RF mean EMG activity when wearing the AE ankle braces. As the quadriceps muscle inserts onto the tibial tuberosity via the patellar ligament, concentric contraction of the quadriceps muscles will cause the tibia to translate anteriorly relative to the femur (DeMorat, Weinhold, Blackburn, Chudik, & Garrett, 2004). This translation will create tension in the ACL, causing external rotation (Tiberio, 1987). Therefore, the small reduction in RF mean EMG activity may create and/or indicate less external rotation and greater knee abduction during forceful knee extension, such as occurs during a vertical jump. In turn, by increasing knee internal rotation and abduction during a vertical jump takeoff, in combination with a neutral or pronated subtalar joint, this may alter the position of the body's centre of mass. Thus, the angle of the GRF vector may have moved laterally, resulting in more force being applied laterally. As the purpose of the vertical jump is to jump as high as possible in the vertical plane, this may indicate an inefficient distribution of force when wearing ankle braces, which was reflected in the reduction of vertical jump height.

Landing

Upon landing, there was a significant decrease in peak lateral GRF. Although there were no other significant findings, there were interesting tendencies in the data.

While not significant, there was a small 6.52 %MVC decrease in PL mean EMG activity at landing when wearing the AE ankle braces. This decrease is similar to the findings of Hopper et al.'s (1999) study, which examined the effects of Swede-o® softshell ankle braces and tape on mean EMG activity of the ankle musculature during a standardized, single leg landing task. Hopper et al. also observed a significant decrease in mean EMG activity of the LG. In the current study, there was a mean decrease of 8.11 %MVC between the no brace and AE conditions, although this decrease was not significant. Furthermore, both studies are in agreement regarding the effects of softshell ankle braces on peak vertical GRF, in that no significant changes were observed between conditions.

The implications of a reduction in the magnitude of the PL contraction when wearing ankle braces has been debated in the literature. As previously stated, the PL muscle may protect the ankle from sudden inversion and/or supination by reflexively contracting and resisting dynamic supination of the ankle joint (Konradsen et al., 1997). As such, if the magnitude of the PL contraction is reduced during a dynamic supination, this natural protective mechanism may be compromised, placing the ankle at greater risk of injury (Cordova et al., 2000). Conversely, the reduction in PL EMG activity may indicate that the ankle brace is absorbing external forces that would otherwise be placed on the musculoskeletal system, thereby reducing the risk on injury (Cordova et al., 1998).

Irrespective of what is happening biomechanically and physiologically at the ankle, ankle braces prevent ankle injury (Leppänen, Aaltonen, Parkkari, Heinonen, & Kujala, 2014; Petersen et al., 2013). The effectiveness, however, of ankle braces relative to other preventative measures may be related to the biomechanical and physiological changes that occur when wearing ankle braces. Although no biomechanical or physiological variables were measured, Stasinopoulos

(2004) found that jump takeoff and landing technique training, as well as proprioceptive training, was more effective in preventing ankle injuries than semi-rigid ankle braces. Furthermore, the Active Ankle T2™ semi-rigid ankle braces proved ineffective in reducing ankle sprains in persons who had incurred more than three ankle sprains, while technique training and proprioceptive training remained effective (Stasinopoulos, 2004). As persons with CAI have been shown to experience decreased activation of the ankle musculature and stabilizers (Santilli et al., 2005), the fact that an active exercise and training program was effective in preventing ankle injuries in this population, while ankle braces were not, supports factors such as the magnitude of the PL contraction at least contributing to ankle sprain prevention. Given that a decrease in PL mean EMG activity was observed in the current study during a maximal vertical jump landing, this may indicate that ankle braces, although effective, should be combined with other prevention strategies.

Many competitive sports teams have policies that require players to wear ankle braces irrespective of their level of ankle function (Pedowitz et al., 2008). When considered with the fact that up to 86% of ankle sprains in volleyball occur when landing from a jump (Bahr et al., 1994), the implications of a reduction in PL mean EMG activity become even more relevant, especially when examining the effects that T1 ankle braces have on vertical GRF in the current study, as well as others. Although Cordova et al. (1998) hypothesized that a reduction in PL EMG activity could indicate that ankle braces were absorbing external forces, peak vertical GRF remained unchanged in the current study when wearing AE or T1 ankle braces. Furthermore, Cordova et al. (2000) observed no change in peak vertical GRF when wearing the McDavid Ultra semi-rigid ankle braces during a standardized drop landing. Hopper et al. (1999) observed no change in peak vertical GRF, alongside a significant reduction in PL EMG activity when wearing the Swede-o® Softshell ankle brace. In comparison, Hodgson et al. (2005) observed a significant increase in peak vertical GRF when wearing the Active Ankle T2™ semi-rigid ankle brace during a standardized drop landing; while Simpson et al. (2013) observed a similar, significant increase in peak vertical GRF when wearing AE ankle braces, although no EMG data were recorded in these studies. As such, it seems unlikely that the decrease in PL EMG activity is the result of ankle braces attenuating the magnitude of GRFs, at least in the vertical plane. As the inability of the musculoskeletal system to absorb GRFs can lead to injury (Dufek & Bates, 1990), and the fact that softshell and semi-rigid ankle braces appear to have no effect and/or increase vertical GRFs during a jump landing, it seems unlikely that ankle braces contribute to injury prevention by attenuating vertical GRFs. When combined with the reduction observed in the magnitude of the PL muscle contraction when wearing softshell ankle braces during landing, this would have negative implications for injury prevention.

While the hypothesis that a decrease in PL EMG activity is indicative of ankle braces absorbing external forces is not supported when examining vertical GRF during landing, it may hold weight when examining peak lateral GRF. Although the decrease in peak lateral GRF was only 2.59 %BW when wearing T1 softshell ankle braces relative to no braces, this difference was significant. As Hodgson et al. (2005) and Hopper et al. (1999) only reported variables related to vertical GRF and kinematics, it is difficult to make comparisons to the current study; however, Simpson et al. (2013) also reported a significant increase in peak lateral GRF, albeit when wearing the AE softshell ankle braces. This increase was accompanied by a reduction in knee internal rotation displacement, knee flexion, and plantarflexion at landing. Again, this decrease in lateral GRF may be due to ankle braces restricting the ankle to a more neutral position, altering the position of the body's centre of mass and the angle at which the GRF vector is applied, as well as the joint kinematics proximal to the ankle.

Although not significant, compared to the no brace condition, total vertical impulse was 6.85 N·s and 6.34 N·s less when wearing the AE and T1 ankle braces, respectively. Since total vertical impulse was not measured in previous studies examining the effect of ankle braces on GRFs during landing, it is difficult to compare the nonsignificant decrease in total vertical impulse observed in the current study when wearing ankle braces. Previous studies, however, have examined the time to peak vertical GRF along with ankle kinematics during landing. As impulse is the integral of force and time (Hamill et al., 2015) combined with kinematic data from other studies, inferences could be made to explain why there was a reduction in total vertical impulse when wearing the T1 ankles braces compared to no braces.

As peak vertical GRF did not increase significantly across brace conditions, the decrease in total vertical impulse could be the result of ankle braces decreasing the average force applied during landing. Alternatively, it may also be the result of ankle braces altering the time over which force is applied. When landing from a vertical jump with two feet, the force-time curve will generally have two peaks; peak one is created from the metatarsal heads impacting the ground after toe contact, and peak two represents contact of the calcaneus with the ground (Abián, Alegre, Lara, Rubio, & Aguado, 2008). As such, landing will normally occur in a "toe to heel" fashion (Bressel & Cronin, 2013). Based on this style of landing, the degree to which the ankle is plantarflexed during toe contact will affect the time to which the second peak is observed; less plantarflexion would create less distance between the heel and the ground when the toe makes contact, therefore, decreasing the time to heel contact after toe contact. Conversely, a larger plantarflexion angle at landing would increase this distance and simultaneously increase the time to heel contact after toe contact (Riemann et al., 2002). As both softshell and semi-rigid ankle braces have been shown to reduce plantarflexion angle and displacement at landing (Cordova et al., 2010; DiStefano et al., 2008; Hodgson et al., 2005; Simpson et al., 2003), and simultaneously decrease the time to peak vertical GRF variables (Cordova et al., 2010; Hodgson et al., 2005; Simpson et al., 2013), it is logical to suggest that the same thing occurred in the current study. As time to peak vertical GRF was not reported for the current study, however, this rationale is speculative. Additionally, because mean GRF was not reported, it is unknown if the average force applied may also have been affected.

An explanation as to why the T1 ankle brace resulted in a decrease in total vertical impulse, while the AE ankle brace did not, may again be found in the design of the T1 ankle brace. As previously discussed, altered cutaneous input as the result of the heel pad of the T1 ankle brace may facilitate the TA muscle, which would promote dorsiflexion, thereby reducing the degree to which the foot was plantarflexed at landing. Therefore, this would have reduced the time to reach peak vertical GRF, contributing to the reduction in total vertical impulse.

As with the decrease observed in PL mean EMG activity during landing when wearing the AE ankle braces, a reduction in total vertical impulse may be interpreted as having a positive, as well as negative effect on injury (Simpson et al., 2013). Negative, in that excessive vertical GRF loading has been suggested to lead to overuse injuries of the lower extremity (Dixon, Collop, & Batt, 2000; Dufek & Bates, 1990; Seegmiller & McCaw, 2003; Yeow, Ng, Lee, & Gow, 2010). Positive, in that larger vertical GRF values and the speed at which vertical GRF is applied is essential for promoting bone growth and strengthening the skeletal system (Nigg, 2001). Therefore, longitudinal studies are required to determine the potential effects of softshell and semi-rigid ankle braces on joint and bone health across the lifespan and at different ages of skeletal maturity (Simpson et al., 2013).

In addition to the potential effect on overuse injuries, the implications of the current study should also be examined within the context of acute knee injuries. The majority of ACL injuries involve a non-contact mechanism, such as landing from a jump or pivoting (Boden, Dean, Feagin, & Garret, 2000; Olsen et al., 2004). When ACL injuries occur during landing, the knee is often in, or close to a fully extended position (Boden et al., 2000). The contraction of the quadriceps muscle associated with greater knee extension angles has been implicated in ACL injuries, in that the quadricep muscles will translate the tibia anteriorly relative to the femur, placing strain on the ACL (Boden et al., 2000; DeMorat, Weinhold, Blackburn, Chudik & Garret, 2004; Draganich & Vahey, 1990). Although nonsignificant, there was a large decrease in RF mean EMG of 13.21 %MVC and 12.31 %MVC when wearing the AE and T1 ankle braces, respectively. DiStefano et al. (2008) found that knee flexion angle increased by 3° at landing when wearing the AE softshell ankle braces, hypothesizing that this increase in knee flexion was a compensatory strategy for the reduction in ankle dorsiflexion. As such, the non-significant decrease in RF mean EMG activity observed across conditions during landing could be indicative of a reduction in knee flexion angle at landing, resulting in decreased strain being placed on the ACL when wearing ankle braces.

Cutting Task

Compared to wearing no ankle brace, there was a significant increase in time to complete the cutting task when wearing ankle braces. This amounted to times that were 0.16 seconds and 0.20 seconds slower when wearing the AE and T1 ankle braces, respectively. As the cutting task was developed specifically for the current study and primarily to elicit a quick change in direction, it is difficult to compare the performance results of the cutting task to previous literature, although EMG characteristics of the task were similar to Colby et al. (2000) and Houck (2003). Therefore, for the purposes of the discussion, comparison will be made to studies utilizing tests that have been validated as measures of agility. Additionally, as a kinematic analysis of wearing ankle braces during an agility test is not present in the literature, comparison will be made to studies examining the kinematics of cutting tasks.

This reduction in performance on a cutting task is comparable to the research of Ambegaonkar et al. (2011), who observed a significant increase in time to complete the Right-Boomerang Run Test when wearing the Aircast Airstirrup™ semi-rigid ankle braces. The results, however, contradict the majority of research that suggests softshell and semi-rigid ankle braces do not affect agility performance. Henderson et al. (2016), Leonard and Rotay (2014), Macpherson et al. (1995), Parsley et al. (2013) all concluded that, in individuals with healthy ankles, softshell and semi-rigid ankle bracing did not affect agility performance. Additionally, all of the studies mentioned came to this conclusion using different tests of agility, indicating that any change in agility performance may be dependent upon the type of test and type of ankle brace used.

Ambegaonkar et al. (2011) hypothesized that the increased time to complete the Right-Boomerang Run Test when wearing ankle braces was due to a restriction in ROM at the ankle; participants were not able to change directions as quickly due to the ankle braces limiting the mobility of the ankle. Based on the results of the current study, however, this explanation is unlikely. During the deceleration and propulsive phases of the cutting task, peak anterior and posterior GRF, and peak medial and lateral GRF were not affected when wearing ankle braces. Additionally, impulse measures were not significantly affected during the deceleration and propulsive phases of the cutting task, indicating that force and time in contact with the force platform remained relatively stable across conditions. Given that the time to complete the cutting task was slower in the AE and T1 conditions, despite no change in GRF variables, ankle braces may be affecting other components of the task. Previous literature examining the effects of ankle braces on kinematics during cutting tasks has focused largely on the change in direction component of the task used (Klem et al., 2016; Greene et al., 2014; West et al., 2013). As agility tests normally involve multiplanar movement, including movement in the sagittal plane, consideration should be given to the effect that ankle braces may have on running in the sagittal plane.

As previously stated, the foot alternates between a pronated and supinated position during gait (Tiberio, 1987), which is anatomically possible due to internal and external rotation of the lower extremity (Inman, 1976; Kirby, 2000). Therefore, a restriction in pronation and supination could result in kinematic changes proximal to the ankle during gait. Although Tamura et al. (2017) did not find any changes in sagittal plane kinematics at the knee or hip during treadmill running when wearing softshell or semi-rigid ankle braces, there was a significant increase in the knee abduction angle at initial foot contact when wearing AE ankle braces and Active Ankle T2™ semi-rigid ankle braces, relative to wearing no braces. This was accompanied by a significant increase in total energy expenditure when wearing the Active Ankle T2™ ankle braces, which had previously been reported by Mackean et al. (1995) when wearing Aircast Airstirrup™ semi-rigid ankle braces. As such, a restriction of frontal plane ROM at the ankle may result in greater displacement of the hip in the frontal plane, creating a less efficient gait pattern when wearing ankle braces. As such, this may have resulted in a slower time to complete the cutting task when wearing ankle braces. Further research is needed, however, to determine the effects of ankle bracing on running and sprinting kinematics, especially the role that fatigue and/or sporting conditions may have on any potential effects.

An important consideration regarding the results of the current study, as well as injury risk, is the relatively large between trial variability of the jump landing and cutting task variables across conditions. Previous literature has identified high trial-to-trial variability for kinematic and EMG variables during jump landings (Fagenbaum & Darling, 2003). Given the heterogenous nature of the population (experienced and non-experienced ankle brace users, multiple types and levels of sport), in addition to an uncontrolled landing height, this variability was to be expected in the current study. This variability, however, may suggest less than optimal landing mechanics and neuromuscular control during jump landings in the study population. Stasinopoulos (2004) suggested that, even in elite level sport, takeoff and landing technique may be sub-par, given the significant decrease in ankle sprains experienced by participants who took part in a technical jump training program. Additionally, Pienkowski et al. (1995) noted that improvements in athletic performance occurred after participants completed a one week adaptation period to softshell and semi-rigid ankle braces. Therefore, based on this information, inexperienced ankle brace users may want to consider an "adjustment period," and/or jump training program, to offset potential changes to lower extremity biomechanics and performance that were observed in the current study when wearing ankle braces.

Limitations

With respect to the interpretation and generalizability of the results of the current study, some limitations must be taken into consideration. As this was an exploratory study, the analysis did not control for independent variables such as sex, ankle bracing experience, or sport, which could influence the dependent variables collected in the current study. Additionally, a univariate analysis was selected over a multivariate analyses, potentially increasing the chance of a type I error.

In terms of the procedures, although the Vertical Jump Test was based on an established protocol, due to the incorporation of a pause, the external validity of the test may be called into question. In an athletic situation, such as basketball or volleyball, a jump would normally incorporate a counterbalance movement to take advantage of the stretch shortening cycle (Hamill et al., 2015), which the Vertical Jump Test removed. Additionally, by constricting the landing to one foot off-one foot on the force platform, this introduced a constraint that would not be present in a real world situation, potentially resulting in the participant adjusting their jump form and impacting on the overall jump height accordingly. This constraint was also present during the lateral cutting task, which required participants to approach the force platform before planting the dominant foot on the force platform. Again, participants may have altered their form, speed, and technique accordingly to remain within the constraints of the task. Furthermore, the results of the current study are only representative of the participant's dominant extremity, again due to participants only contacting the platform with the dominant foot, as well as only having electrodes on the dominant lower extremity.

As the three trials for each condition were averaged for each variable, the high variability between trials for EMG and kinetic variables during the landing phase and cutting task may have attenuated extreme values, improving the validity of each value. The Vertical Jump Test and cutting task emphasized performance and were very reliable in terms of measuring vertical jump height and time to complete the cutting task. Form for the vertical jump was standardized for the takeoff phase, likely resulting in the high trial to trial reliability for all measures during the takeoff phase. Due to this emphasis on performance, however, variables such as fatigue, individual landing technique, and participants' proficiency with the movements may have contributed to the variability in EMG and kinetic measures from trial to trial during the landing phase and cutting task. Furthermore, as variables such as height and speed were not controlled, any fluctuation in these variables would likely have influenced EMG and kinetic measures.

Delimitations

Delimitations of the study must also be taken into consideration. Rather than using a homogeneous, one sport population, a heterogenous population incorporating a wide variety of sports, as well as ankle bracing experience was selected. This was done in an effort to maximize sample size. As such, the results of the current study may not be entirely representative of a single sport or athletic population, but may have implications in a broader sport context. Although analyzing the maximum or median value of the three trials would have limited the influence of high trial to trial variability on the results, the mean value of each variable was used in the interest of external validity, as over the course of a game or practice, athletes may make movements, such as a jump or cut many times (Bompa & Haff, 2009), potentially resulting in fatigue and decreased performance.

For the analysis, a univariate analysis was selected over a multivariate analysis due to the research questions, exploratory nature of the study, and lack of comparable studies or models. This also allowed data for some variables to remain in the analysis, despite their being missing data for other variables. Furthermore, running a MANOVA prior to an ANOVA does not necessarily control for type 1 error, due to the alpha level of the ANOVA only being less than or equal to the alpha level for the MANOVA when the MANOVA's null hypothesis is accepted (Bird & Hadzi-Pavlovic, 1983). Rather, a Bonferroni adjustment was used to control for type 1 error.

Future Research

As the current study was the first to incorporate EMG of proximal lower extremity musculature when assessing the effects of ankle braces during athletic performance measures, more research is needed to support the results of the current study. Further research should incorporate kinematic analysis to improve the interpretation of the EMG and kinetic data. Additionally, future research should consider using a homogenous athletic population, as well as a pathological population, such as CAI, as ankle brace use is not limited to healthy individuals.

Although not directly analyzed as part of the current study, sex differences in muscular activation during landing should also be addressed. Female athletes are 1.7 times more likely to sustain an ACL injury than male athletes (Montalvo et al., 2018). Many reasons for this difference have been suggested, including an overreliance on the quadriceps muscles by females to stabilize the knee joint (Hewett, Ford, Hoogenbook, & Myer, 2010; Hewett, Stroupe, Nance, & Noyes, 1996). As such, females also display lower hamstring activation relative to males when landing from a jump (Urabe et al., 1999). Descriptively, females displayed similar RF mean EMG activity to male counter parts in the no brace control condition during landing, but greater RF mean EMG activity in the AE condition. Additionally, although males displayed much higher BF mean EMG activity in the no brace control and T1 conditions during landing, the BF mean EMG activity was almost identical in the T1 condition. Given the differences in injury rates between males and females when wearing ankle braces (Frey et al. 2010), further research should consider the role that sex may have on ankle bracing's effects on lower extremity biomechanics.

Chapter Six: Conclusion

The purpose of the current study was to determine if wearing semi-rigid and softshell ankle braces affects vertical jump height and time to complete a lateral cutting task, as well as kinetics and lower extremity EMG activity during these measures. The current study builds on and supports previous work examining the effects of softshell and semi-rigid ankles braces on vertical jump height and agility tasks, and adds to the limited literature examining lower extremity EMG and kinetics during athletic performance tasks.

The results of the current study are in line with previous studies, in that vertical jump height was significantly decreased by wearing softshell and semi-rigid ankle braces. Furthermore, semi-rigid ankle braces significantly decreased mean EMG activity of the LG muscle during the takeoff phase of the Vertical Jump Test. Additionally, softshell and semi-rigid ankle braces significantly increased peak lateral GRF during the takeoff phase. During the landing phase, semi-rigid ankle braces significantly decreased peak lateral GRF. Softshell and semi-rigid ankle braces significantly increased time to complete a cutting task, but did no significantly affect any EMG or kinetic variables during the cutting task.

Based on the findings of the current study, softshell and semi-rigid ankle braces decrease vertical jump height and cutting performance. Softshell and semi-rigid ankle braces may also influence the EMG activity and kinetic variables associated with lower extremity injury during a vertical jump landing, although this is still unclear. Any effects on EMG or kinetics, however, may be dependent on brace design and function. Further, more comprehensive research is needed incorporating kinematic, kinetic, and EMG analysis to determine the exact mechanisms behind the observed decrease in performance, and potential implications for injury and injury prevention. In the mean time, clinicians, athletes, trainers, and any users or prescribers of ankle braces should weigh the pros and cons of prophylactically bracing the ankle, especially from a performance perspective.

References

Abián, J; Alegre, L M; Lara, A J; Rubio, J A; Aguado, X. (2008). Landing differences between men and women in a maximal vertical jump amplitude test. *Journal of Sports Medicine and Physical Fitness, 48*, 305-310.

Alfuth, M., Klein, D., Koch, R., & Rosenbaum, D. (2014). Biomechanical comparison of 3 ankle braces with and without free rotation in the sagittal plane. *Journal of Athletic Training, 49*, 608-616.

Advanced Medical Technologies, Inc. (AMTI). (1987). *Biomechanics platform set instruction manual*. Newton, MA.

Anderson, M. K., & Parr, G. P. (2013). *Foundations of athletic training* (5th ed.). Baltimore & Philadelphia, U.S.A: Lippincott Williams & Wilkins.

Ambegaonkar, J. P., Redmond, C. J., Winter, C., Ambegaonkar, S. J., Thompson, B., & Guyer, S. M. (2011). Ankle stabilizers affect agility but not vertical jump or dynamic balance performance. *Foot and Ankle Specialist, 4*, 354-360.

Aragón-Vargas, L. F. (2002). Evaluation of four vertical jump tests: Methodology, reliability, validity, and accuracy. *Measurement in Physical Education and Exercise Science, 4*, 215-228. doi: http://dx.doi.org/10.1207/S15327841MPEE0404_2

Aragón-Vargas, L. F., & Gross, M. M. (1997). Kinesiological factors in vertical jump performance: Differences among individuals. *Journal of Applied Biomechanics, 13*, 24-44.

Bachman, L. M., Kolb, E., Koller, M. T., Steurer, J., ter Riet, G. (2003). Accuracy of Ottawa ankle rules to exclude fractures of the ankle and mid-foot: a systematic review. *British Medical Journal, 326*, 417-424. https://doi.org/10.1136/bmj.326.7386.417

Bahr, R., & Bahr, I. A. (1997) Incidence of acute volleyball injuries: a prospective cohort study of injury mechanisms and risk factors. *Scandinavian Journal of Medicine and Science in Sports, 7*, 166-171. doi: 10.1111/j.1600-0838.1997.tb00134.x

Bahr, R., Karlsen, R., Liam, Ø., & Øvrebø. (1994). Incidence and mechanisms of acute ankle inversion injuries in volleyball. *American Journal of Sports Medicine, 22*, 595-600.

Bahr, R., Lian, O., & Bahr, I. A. (1997). A twofold reduction in the incidence of ankle sprains in volleyball after introduction of an injury prevention program: A prospective cohort study. *Scandinavian Journal of Medicine and Science in Sports, 7*, 172-177.

Barlow, G., Donovan, L., Hart, J. M., & Hertel, J. (2015). Effect of lace-up braces on electromyography measures during walking in adults with chronic ankle instability. *Physical Therapy in Sport, 16*, 16-21. https://doi.org/10.1016/j.ptsp.2014.02.002

van den Bekerom, M. P. J., Oostra, R. J., Golanó, P., & van Dijk, C. N. (2008). The anatomy in relation to injury of the lateral collateral ligaments of the ankle: A current concepts review. *Clinical Anatomy, 21*, 619-626. doi: 10.1002/ca.20703

Beynnon, B. D., Murphy, D. F., & Alosa, D. M. (2002). Predictive factors for lateral ankle sprains: a literature review. *Journal of Athletic Training, 37*, 376-380.

Bird, K. D. & Hadzi-Pavlovic, D. (1983). Simultaneous test procedures and the choice of a test statistic in MANOVA. *Psychological Health Bulletin, 93,* 167-178.

Boardman, D. L., & Liu, S. H. (1997). Contribution of the anterolateral joint capsule to the mechanical stability of the Ankle. *Clinical Orthopaedics and Related Research, 341*, 224-232.

Boden, B. P., Dean, S. G., Feagin, J. A. Jr., & Garret, W. E. (2000). Mechanisms of anterior cruciate ligament injury. *Orthopedics, 23,* 573-578. doi: 10.3928/0147-7447-20000601-15

Bompa, T. O., & Haff, G. G. (2009). *Periodization: Theory and methodology of training.* Champaign, IL: Human Kinetics.

Bot, S. D. M., & van Mechelen, W. (1999). The effect of ankle bracing on athletic performance. *Sports Medicine, 27*, 171-178.

Bressel, E., & Cronin, J. (2013). The landing phase of a jump strategy to minimize injuries. *Journal of Physical Education, Recreation, and Dance, 76*(2), 20-35. https://doi.org/10.1080/07303084.2005.10607332

Bullock-Saxton, J. E., Janda, V., & Bullock, M. I. (1994). The influence of ankle sprain injury on muscle activation during hip extension. *International Journal of Sports Medicine, 15*, 330-334.

Burks, R. T., Bean, B. G., Marcus, R., & Barker, H. B. (1991). Analysis of athletic performance with prophylactic ankle devices. *American Journal of Sports Medicine, 12*, 104-106. doi: 10.1177/036354659101900202

de Comargo Neves Sacco, I., Takahasi, H. Y., Suda, E. Y., Battistella, L. R., Kavamoto, C. A., Lopes, J. A. A., & de Vasconcelos, J. C. P. (2006). Ground reaction force in basketball cutting maneuvers with and without ankle bracing and taping. *Sao Paulo Medical Journal. 124,* 245-252.

Cloak, R., Galloway, S., & Wyon, M. (2010). The effect of ankle bracing on peak mediolateral ground reaction force during cutting maneuvers in collegiate male basketball players. *Journal of Strength and Conditioning Research, 24*, 2429-2433. doi: 10.1519/JSC.0b013e3181e2e0d5

Colby, S., Francisco, A., Yu, B., Kirkendall, D., Finch, M., & Garret, W. (2000). Electromyographic and kinematic analysis of cutting maneuvers: Implications for anterior cruciate ligament injury. *American Journal of Sports Medicine, 28*, 234-240.

Cordova, M. L., Armstrong, L., Rankin, C. W., James, M., & Yeasting, R. A. (1998). Ground reaction forces and emg activity with ankle bracing during inversion stress. *Medicine and Science in Sport and Exercise. 30*, 1363-1370.

Cordova., M. L., & Armstrong, L. (1996). Reliability of ground reaction forces during a vertical jump: implications for functional strength assessment. *Journal of Athletic Training, 31*, 342-345.

Cordova, M. L., Cardona, C. V., Ingersoll, C. D., & Sandrey, M. A. (2000). Long-term ankle brace use does not affect peroneus longus muscle latency during sudden inversion in normal subjects. *Journal of Athletic Training, 35*, 407-411.

Cordova, M. L., & Ingersoll, C. D. (2003). Peroneus longus stretch reflex amplitude increases after ankle brace application. *British Journal of Sports Medicine, 37*, 258-262. doi:10.1136/bjsm.37.3.258

Cordova, M., Scott, B. D., Ingersoll, C. D., & LeBlanc, M. J. (2005). Effects of ankle support on lower-extremity functional performance: a meta-analysis. *Medicine and Science in Sports and Exercise, 37*, 635-641.

Cordova, M. L., Takahashi, Y., Kress, G. M., Brucker, J. B., & Finch, A. E. (2010). Influence of external ankle support on lower extremity joint mechanics during drop landings. *Journal of Sport Rehabilitation, 19*, 136-148.

Criswell, E. (2011). Introduction to surface electromyography (2nd ed.). Sudbury, USA: Jones and Bartlett Publishers.

Cronin, J. B., & Templeton, R. L. (2008). Timing light height affects sprint times. *Journal of Strength and Conditioning Research, 22*, 318-320. doi: 10.1519/JSC.0b013e31815fa3d3.

CSEP. (2013). *Canadian society for exercise physiology- physical activity training for health*. Ottawa, Canada: Canadian Society for Exercise Physiology.

Delsys Inc.. (2012). *Delsys Trigno Wireless system: User's guide*. Boston, MA: Delsys Inc.

DeMorat, G., Weinhold, P., Blackburn, T., Chudik, S., & Garret, W. (2004). Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *American Journal of Sports Medicine, 32*, 477-483. doi: 10.1177/0363546503258928

Devita, P., & Skelly, W. A. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Medicine and Science in Sports and Exercise, 24,* 108-115.

van Dijk, C. N. (1994). *On diagnostic strategies in patient with severe ankle sprain* (Thesis). University of Amsterdam.

van Dijk, C. N., Bossuyt, P. M. M., & Marti, R. K. Medial ankle pain after lateral ligament rupture. *Journal of Bone and Joint Surgery, 78*, 562-564.

DiStefano, L. J. Padua, D. A., Brown, C. N., & Guskiewicz, K. M. (2008). Lower extremity kinematics and ground reaction forces after prophylactic lace-up ankle bracing. *Journal of Athletic Training, 43*, 234-241. doi: 10.4085/1062-6050-43.3.234.

Dixon, S. J., Collop, A. C., & Batt, M. E. (2000). Surface effects on ground reaction forces and lower extremity kinematics in running. *Medicine and Science in Sports and Exercise, 32*, 1921-1926.

Donatelli, R. (1987). Abnormal biomechanics of the foot and ankle. *Journal of Orthopedic and Sports Physical Therapy, 9*, 11-16.

Donnell-Fink, L. A., Klara, K., Collins, J. E., Yang, H. Y., Goczalk, M. G., Natz, J. N., & Losina, E. Effectiveness of knee injury and anterior cruciate ligament tear prevention programs: a meta-analysis. *PLOS ONE, 10*(12). 1-17. https://doi.org/10.1371/journal.pone.0144063

Draganich, L. F., Jaeger, R. J., Kralj, A. R. (1989). Coactivation of the hamstrings and quadriceps during extension of the knee. *Journal of Bone and Joint Surgery, 71a*, 1075-1081.

Draganich, L. F., & Vahey, J. W. (1990). An in vitro study of anterior cruciate ligament strain induced by quadriceps and hamstrings forces. *Journal of Orthopaedic Research, 8*, 57-61. https://doi.org/10.1002/jor.1100080107

Drewes, L. K., McKeon, P. O., Kerrigan, C., & Hertel, J. (2009). Dorsiflexion deficit during jogging with chronic ankle instability. *Journal of Science and Medicine in Sport, 12*, 685-687. DOI: https://doi.org/10.1016/j.jsams.2008.07.003

Dufek, J. S., & Bates, B. T. (1990). The evaluation and prediction of impact forces during landings. *Medicine and Science in Sports and Exercise, 22*, 370-377.

Dutton, M. (2012). *Dutton's orthopaedic examination, evaluation, and intervention* (3rd ed.). United States: McGraw-Hill.

Eachaute, C., Vaes, P., Van Aerschot, L., Asman, S., & Duquet, W. (2007). The clinometric qualities of patient-assessed instruments for measuring chronic ankle instability: A systematic review. *BMC Musculoskeletal Disorders, 8*(6). doi: 10.1186/1471-2474-8-6

Eils, E., Demming, C., Kollmeier, G., Thorwesten, L., Völker, K., & Rosenbaum, D. (2002). Comprehensive testing of 10 different ankle braces. Evaluation of passive and rapidly induced stability in subjects with chronic ankle instability. *Clinical Biomechanics, 17*, 526-535. http://dx.doi.org/10.1016/S0268-0033(02)00066-9

Fallat, L., Grimm, D. J., & Saracco, J. A. (1998). Sprained ankle syndrome: Prevalence and analysis of 639 acute injuries. *Journal of Foot and Ankle Surgery, 37*, 280-285.

Fagenbaum, R., & Darling, W. G. (2003). Jump landing strategies in male and female college athletes and implications of such strategies for anterior cruciate ligament injury. *American Journal of Sports Medicine, 31, 233-240*. doi: 10.1177/03635465030310021301

Feuerbach, J. W., Grabiner, M. D., Koh, T. J., & Weiker, G. G. (1994). Effect of an ankle orthosis and ankle ligament anesthesia on ankle joint proprioception. *American Journal of Sports Medicine, 22*, 223-229. doi: 10.1177/036354659402200212

Feger, M. A., Donovan, L., Hart, J. M., & Hertel, J. (2014). Lower extremity muscle activation during functional exercises in patients with and without chronic ankle instability. *PM & R: The Journal of Injury, Function, and Rehabilitation, 6*, 602-611. doi: 10.1016/j.pmrj.2013.12.013

Feger, M. A., Donovan, L., Hart, J. M., & Hertel, J. (2015). Lower extremity muscle activation in patients with or without chronic ankle instability during walking. *Journal of Athletic Training, 50*, 350-357. doi: 10.4085/1062-6050-50.2.06

Ferran, N. A., Maffulli, N. (2006). Epidemiology of sprains of the lateral ankle ligament complex. *Foot and Ankle Clinics, 11*, 659-662. http://dx.doi.org/10.1016/j.fcl.2006.07.002

Fauth, M., Petushek, E., Feldmann, C., Hsu, B., & Garceau, L. (2010). Reliability of surface electromyography during maximal voluntary isometric contractions, jump landings, and cutting. *Journal of Strength and Conditioning Research, 24*, 1131-1137. doi:10.1519/JSC.0b013e3181cc2353

Freeman, M. A. R. (1965). Instability of the foot after injuries to the lateral ligament of the ankle. *Journal of Bone and Joint Surgery, 47B*, 669-677.

Freeman, M. A. R., Dean, M. R. E., & Hanham, I. W. F. (1965). The etiology and prevention of functional instability of the foot. *Journal of Bone and Joint Surgery, 47B*, 678-685.

Frey, C., Feder, K. S., & Sleight, J. (2010). Prophylactic ankle brace use in high school volleyball players: A prospective study. *Foot and Ankle International, 31*, 296-300. doi:10.3113/FA1.2010.0296

Green, T., Refshauge, K., Crosbie, J., & Adams, R. (2001). A randomized controlled trial of a passive accessory joint mobilization on acute ankle inversion sprains. *Physical Therapy, 81*, 984-994.

Greene, T. A., & Roland, C. (1989). A comparative isokinetic evaluation of a functional ankle orthosis on talocalcaneal function. *Journal of Orthopaedic and Sports Physical Therapy, 11*, 245-252.

Greene, A. J., Stuelcken, M. C., Smith, R. M., & Vanwanseele, B. (2014). The effect of external ankle support on the kinematics and kinetics of the lower limb during a side step cutting task in netballers. *BMC Sports Science, Medicine, and Rehabilitation, 6*(42). 1-10. doi: 10.1186/2052-1847-6-42

Gribble, P. A., Hertel, J , & Denegar, C. R . (2007). Chronic ankle instability and fatigue create proximal joint alterations during performance of the Star Excursion Balance Test. *International Journal of Sports Medicine,* 28, 236–242. doi: 10.1055/s-2006-924289

Gribble, P. A., Radel, S., & Armstrong, C. W. (2006) The effects of ankle bracing on the activation of the peroneal muscles during a lateral shuffling movement. *Physical Therapy in Sport, 7*(1), 14-21. http://dx.doi.org/10.1016/j.ptsp.2005.10.003

Gribble, P. A., & Robinson, R. H. (2009). Alteration in knee kinematics and dynamic stability associated with chronic ankle instability. *Journal of Athletic Training, 44*, 350-355.

Gross, M. T., Bradshaw, M. K., Ventry, L. C., & Weller, K. H. (1987). Comparison of support provided by ankle taping and semirigid orthosis. *Journal of Orthopaedic and Sports Physical Therapy, 9*, 33-39. doi: 10.2519/jospt.1987.9.1.33

Gross, T. S., & Nelson, R. C. (1988). The shock attenuation role of the ankle during landing from a vertical jump. *Medicine and Science in Sports and Exercise, 20*, 506-514.

Gudibanda, A., & Wang, Y. (2005). Effect of the ankle stabilizing orthosis on foot and ankle kinematics during cutting maneuvers. *Research in Sports Medicine, 13*(2), 111-126. doi:10.1080/15438620590956115

Halaki, M., & Ginn, K. (2012). Normalization of EMG signals: To normalize or not to normalize and what to normalize to? In G. R. Naik (Ed.), *Computational Intelligence in Electromyography Analysis - A Perspective on Current Applications and Future Challenges* (175-194). InTech. doi: 10.5772/49957

Hall, S. J. (2012). *Basic biomechanics* (6th ed.). New York, New York: McGraw-Hill.

Hamill, J., Knutzen, K. M., & Derrick, T. R. (2015). *Biomechanical basis of human movement* (4th ed.). Philadelphia, U.S.A.: Lippincott Williams & Wilkens.

Heit, E. J., Lephart, S. M., & Rozzi, S. L. (1996). The effect of ankle bracing and taping on joint position sense in the stable ankle. *Journal of Sport Rehabilitation, 5*, 206-213. http://dx.doi.org/10.1123/jsr.5.3.206

Henderson, Z., & Sanzo, P. (2017). *The Biomechanical Effects of Ankle Bracing During Performance Measures ─ A Pilot Study.* Unpublished manuscript, School of Kinesiology, Lakehead University, Thunder Bay, Canada.

Henderson, Z., Sanzo, P., & Zerpa, C. (2016). The effect of prophylactic ankle bracing on physical performance measures in jumping athletes. *International Journal of Sports Science, 6*, 138-145. doi: 10.5923/j.sports.20160604.02

Hertel, J. (2000). Functional instability following lateral ankle sprain. *Sports Medicine, 29*, 161-371.

Hertel, J. (2002). Functional anatomy, pathomechanics, and pathophysiology of lateral ankle instability. *Journal of Athletic Training, 37*, 364-375.

Hewett, T. E., Ford, K. R., Hoogenboom, B. J., & Myer, G. D. (2010). Understanding and preventing ACL injuries: Current biomechanical and epidemiological considerations – update 2010. *North American Journal of Sports Physical Therapy, 5*, 234-251.

Hewett, T. E., Lindenfield, T. N., Riccobene, J. V., & Noyes, F. R. (1999). The effect of neuromuscular training on the incidence of knee injury in female athletes: a prospective study. *American Journal of Sports Medicine, 27*, 699-706. DOI: https://doi.org/10.1177/03635465990270060301

Hewett, T. E., Stroupe, A. L., Nance, T. A., & Noyes, F. R. (1996). Plyometric training in female athletes. Decreased impact forces and increased hamstring torques. *American Journal of Sports Medicine, 27,* 699-706.

Hewett, T. E., Torg, J. S., & Boden, B. P. (2009). Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: Lateral trunk and knee abduction motion are combined components of the injury mechanism. *British Journal of Sports Medicine, 43*, 417-422.

Hoch, M. C., Staton, G. S., Medina Mckeon, J. M., Mattacola, C. G., & McKeon, O. (2012). Dorsiflexion and dynamic postural control deficits are present in those with chronic ankle instability. *Journal of Science and Medicine in Sport, 15*, 574-579. https://doi.org/10.1016/j.jsams.2012.02.009

Hodgson, B., Tis, L., Cobb, S., & Higbie, E. (2005). The effect of external ankle support on vertical ground-reaction force and lower body kinematics. *Journal of Sport Rehabilitation, 14*, 301-312. http://dx.doi.org/10.1123/jsr.14.4.301

Holmer, P., Sondergaard, L., Konradsen, L., Nielson, P. T., & Jorgensen, L. N. (1994). Epidemiology of sprains in the lateral ankle and foot. *Foot and Ankle International, 15*, 72-74.

Hopper, D. M., McNair., P., & Elliot, B. C. (1999) Landing in netball: Effects of taping and bracing the ankle. *British Journal of Sports Medicine. 33*, 409-413. doi:10.1136/bjsm.33.6.409

Houck, J. (2003). Muscle activation patterns of selected lower extremity muscles during stepping and cutting tasks. *Journal of Electromyography and Kinesiology, 13*, 545-554. http://dx.doi.org/10.1016/S1050-6411(03)00056-7

Hsu, W. L., Krishnamoorthy, V., Scholz, J. P. (2009). An alternative test of electromyographic normalization in patients. *Muscle and Nerve, 33*, 232-241. https://dx.doi.org/10.1002/mus.20458

Hubbard, T. J., & Hertel, J. (2008). Anterior positional fault of the fibular after sub-acute lateral ankle sprains. *Manual Therapy, 13*(1). 63-67. doi: 10.1016/j.math.2006.09.008

Jastifer, J. R., & Gustafson, P. A. (2014). The subtalar joint: Biomechanics and functional representations in the literature. *The Foot, 24*, 203-209. http://dx.doi.org/10.1016/j.foot.2014.06.003

Inman, V. T. (1976). *The joints of the ankle.* Baltimore: Williams & Wilkins.

Johnson, M. B., & Johnson, C. L. (1993). Electromyographic response of peroneal muscles in surgical and nonsurgical injured ankles during sudden inversion. *Journal of Orthopaedic and Sports Physical Therapy, 18*, 497-501.

Johnson, E. O., & Soucacos, P. N. (2010). Proprioception. In *International Encyclopedia of Rehabilitation.*

Jones, D. C. (1993). Tendon disorders of the foot and ankle. *Journal of the American Academy of Orthopaedic Surgeons, 1*, 87-94.

Kamen, G., & Gabriel, D. A. (2010). *Essentials of electromyography.* Windsor: Human Kinetics.

Kaminski, T.W., Hertel, J., Amendoloa, N., Docherty, C. L., Dolan, M. G. Hopkins, J. T., Nussbaum, E., Poppy, W., & Richie, D. (2013). National athletic trainers association position statement: Conservative management and prevention of ankle sprains in athletes. *Journal of Athletic Training, 48*, 528-545. doi: 10.4085/1062-6050-48.02

Kennedy, P. M., & Inglis, T. J. (2002). Distribution and behaviour of glabrous cutaneous receptors in the human foot sole. *Journal of Physiology, 538,* 995-1002. DOI: 10.1013/jphysiol.2001.013087

Kirby, K. A. (2000). Biomechanics off the normal and abnormal foot. *Journal of the American Podiatric Medical Association. 90*(1), 30-34. DOI: 10.7547/87507315-90-1-30

Kleinrensink, G. J., Stoeckart, R., Meulstee, J., Kaulesar Sukul, D. M., Vleeming, A., Snijders, C. J., & van Noort, A.(1994). Lowered motor conduction velocity of the peroneal nerve after inversion trauma. *Medicine and Science in Sports and Exercise, 26*, 877-883.

Klem, N-R., Wild, C. Y., Williams, S. A., & Ng, L. (2016). Effect of external ankle support on ankle and knee biomechanics during the cutting maneuver in basketball players. *American Journal of Sports Medicine*. Advance online publication. doi: 10.1177/0363546516673988

Konrad, K. (2005). *The ABC of EMG: a practical introduction to kinesiological electromyography*. USA: Noraxon.

Konradsen, L. (2002). Factors contributing to chronic ankle instability: kinesthesia and joint position sense. *Journal of Athletic Training*, *37*, 381–385.

Konradsen, L., & Ravn, J. B. (1990). Ankle instability caused by prolonged peroneal reaction time. *Acta Orthopaedica Scandinavica, 61*, 388-390.

Konradsen, L., & Ravn, J. B. (1991). Prolonged peroneal reaction time in ankle instability. *International Journal of Sports Medicine, 12*, 290-292. doi: 10.1055/s-2007-1024683

Koo, T. K., & Li, M. Y. (2016). A guideline of selecting and reporting intraclass correlation coefficients for reliability research. *Journal of Chiropractic Medicine, 15*, 155-163. DOI: 10.1016/j.jcm.2016.02.012

Leonard, T. & Rotay, J. (2014). Effects of ankle taping and bracing on agility, vertical jump, and power. *Keystone Journal of Undergraduate Research, 2*(1), 23-28.

Lephart, S. M., Pincivero, D. M., & Rozzi, S. L. (1998). Proprioception of the ankle and knee. *Sports Medicine, 25*, 149-155. https://doi.org/10.2165/00007256-199825030-00002

Leppänen, M., Aaltonen, S., Parkkari, J., Heinonen, A., & Kujala, U. M. (2014). Interventions to prevent sports related injuries: A systematic review and meta-analysis of randomized controlled trials. *Sports Medicine,44*, 473-486. doi: 11007/s40279-013-0136-8

MacKean, L. C., Bell, G., & Burnham, R. S. (1995). Prophylactic ankle bracing vs. taping: effects on functional performance in female basketball players. *Journal of Orthopaedic and Sports Physical Therapy, 22*(2), 77-81.

Macpherson, K., Sitler, M., Kimura, I., & Horodyski, M. (1995). Effects of a semirigid and softshell prophylactic ankle stabilizer on selected performance tests among high school football players. *Journal of Orthopaedic & Sports Physical Therapy, 21*(3), 147-152.

Macrum, E., Bell, D. R., Boling, M., Lewek, M., & Padua, D. (2012). Effect of limiting ankle-dorsiflexion range of motion on lower extremity kinematics and muscle-activation patterns during a squat. *Journal of Sport Rehabilitation*, *21*, 144-150.

Magee, D. J. (2014). *Orthopedic physical assessment* (6th ed.). St. Louis, Missouri: Elsevier Saunders.

Magnusson, M., Enbom, H., Johansson, R. & Pyykkö, I. (1990). Significance of pressure input from the human feet in anterior-posterior postural control. *Acta Otolaryngol, 110,* 182-188.

Malinzak, R., Colby, S. M., Kirkendall, D. T., Yu, B., & Garret, W. E. A. (2001). Comparison of knee joint motion patterns between men and women in selected athletic tasks. *Clinical Biomechanics, 16*, 438-445.

Man, I. O., Morrissey, M. C., & Cywinski, J. K. (2007). Effect of neuromuscular electrical stimulation on ankle swelling in the early period of ankle sprain. *Physical Therapy, 87*, 53-65. http://dx.doi.org/10.2522/ptj.20050244

Martin, R. L., Davenport, T. E., Paulseth, S., Wukich, D. K., & Godges, J. J. (2013). Ankle stability and movement coordination impairments: Ankle ligament sprains. *Journal of Orthopaedic and Sports Physical Therapy, 43*(9), A1-A40. doi: 10.2519/jospt.2013.0305.

Mavi, A., Yildirim, H., Grunes, T., Pestamalci, E., & Gumusburun, E. (2002). The fibular incisura of the tibia with recurrent sprained ankle on magnetic resonance imaging. *Saudi Medical Journal, 23*, 845-849.

Mazieres, B., Rouanet, S., Velicy, J., Scarsi, C., & Reiner, V. (2005). Topical ketoprofen patch (100 mg) for the treatment of ankle sprain a randomized, double-blind, placebo-controlled study. *American Journal of Sports Medicine, 33*, 551-523. doi: 10.1177/0363546504268135

McGuine, T. A., Brooks, A., & Hetzel, S. (2011). The effect of lace-up ankle braces on injury rates in high school basketball players. *American Journal of Sports Medicine, 39*, 1840-1848. doi: 10.1177/0363546511406242

McGuine, T. A., Hetzel., S., Wilson, J., & Brooks, A. (2012). The effect of lace-up ankle braces on injury rates in high school football players. *American Journal of Sports Medicine, 40*, 49-57. doi: 10.1177/0363546511422332

McKay, G. D., Goldie, P. A., Payne, W. R. & Oakes. (2001). Ankle injuries in basketball: injury rate and risk factors. *British Journal of Sports Medicine, 35*, 103-108. doi:10.1136/bjsm.35.2.103

McNair, P. J., & Prapavessis, H. (1999). Normative data of vertical ground reaction forces during landing from a jump. *Journal of Science and Medicine in Sport, 2*, 86-88. https://doi.org/10.1016/S1440-2440(99)80187-X

Merletti, R. D., Farina, H. J., Hermens, H. J., Freriks, B., & Harlaar, J. (1999). European recommendations for signal processing methods for surface electromyography. In H. J. Hermens, B. Freriks, R. Merletti, D. F. Stegeman, J. H. Block, G. Rau., C. Disselhorst-Klug, & G. Hagg (Eds.), *European recommendations for surface electromyography* (57-70). Enschede, Netherlands: Roessingh Research and Development.

Miller, H., Needle, A. R., Swanik, B., Gustavsen, G. A., & Kaminski, T. W. (2012). Role of external prophylactic support in restricting accessory ankle motion after exercise. *Foot and Ankle International, 33*, 862-869. https://doi.org/10.3113/FAI.2012.0862

Montalvo, A. M., Schneider, D. K., Yut, L., Webster, K. E., Beynnon, B., Kocher, M. S., & Myer, G. D. (2018). "What's my risk of sustaining an ACL injury while playing sports?" A systematic review with meta-analysis. *British Journal of Sports Medicine,* Advance online publication. DOI: 10.1136/bjsports-2016-096274

Mulligan, B. (1995). *Manual therapy: "NAGS", "SNAGS", "MWMS" etc.* (3rd ed. ed.) Wellington: Plane View Services Ltd.

Munro, C. F., (1987). Ground reaction forces in running: a reexamination. *Journal of Biomechanics, 20*, 147-155. https://doi.org/10.1016/0021-9290(87)90306-X

Nilsson, J., & Thorstensson, A. (1989). Ground reaction forces at different speeds of human walking and running. *Acta Physiologica, 136*, 217-227. doi: 10.1111/j.1748-1716.1989.tb08655.x

Nigg, B. M. (2001). The role of impact forces and foot pronation: a new paradigm. *Clinical Journal of Sports Medicine, 11*(1), 2-9. 2–9. doi:10.1097/00042752-200101000-00002

Nitz, A. J., Dobner, J. J., & Kersey, D. (1985). Nerve injury and grades II and III ankle sprains. *American Journal of Sports Medicine, 13*, 177-182. https://doi.org/10.1177/036354658501300306

Noyes, F. R., Mooar, P. A., Matthews, D. S., & Butler, D. L. (1983). The symptomatic anterior cruciate-deficient knee. Part 1: The long-term functional disability in athletically active individuals. *Journal of Bone and Joint Surgery, 65*, 154-162.

Noyes, F. R., Mooar, P. A., Matthews, D. S., & Butler, D. L. The symptomatic anterior cruciate-deficient knee. Part 1: The long-term functional disability in athletically active individuals. *Journal of Bone and Joint Surgery, 65*, 154-162.

Nuzzo, J. L., Anning, J. H., & Scharfenberg, J. M. (2011). The reliability of three devices used for measuring vertical jump height. *Journal of Strength and Conditioning Research, 25*, 2580-2590. doi: 10.1519/JSC.0b013e3181fee650

Nurse, M. A., & Nigg B. M. (2001). The effect of changes in foot sensation on plantar pressure and muscle activity. *Clinical Biomechanics, 16*, 719-727. https://doi.org/10.1016/S0268-0033(01)00090-0

Olmsted, L. C., Vela, L. I., Denegar, C. R. & Hertel, J. (2004). Prophylactic ankle taping and bracing: A numbers-needed-to-treat and cost-benefit analysis. *Journal of Athletic Training, 39*, 95-100.

Olsen, O. E., Myklebust, G., Engebretsen, L., & Bahr, R. (2004). Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *American Journal of Sports Medicine, 32*, 1002-1012. doi: 10.1177/0363546503261724

Pandy, M. G., & Zajac, F. E. (1991). Optimal muscular coordination strategies for jumping. *Journal of Biomechanics, 24*(1), 1-10. https://doi.org/10.1016/0021-9290(91)90321-D

Paris, D. L. (1992). The effects of the Swedo-o, New Cross, and McDavid ankle braces and adhesive ankle taping on speed, balance, agility, and vertical jump. *Journal of Athletic Training, 27*, 253-256.

Parsley, A., Chinn, L., Lee, S. Y., Ingersoll, C., & Hertel, J. (2013). Effect of 3 different ankle braces on functional performance and ankle range of motion. *Athletic Training and Sports Health Care, 5*(2), 69-75.

Paule, K., Madole, K., Garhammer, J., Lacourse, M., & Rozenek, R. (2000). The reliability and validity of the t-test as a measure of agility, leg power, and leg speed in college-aged men and women. *Journal of Strength and Conditioning Research, 14*(4), 443-450.

Pedowitz., D. I., Reddy, S., Parekh, S. G., Huffman, R., & Sennet, B. J. (2008). Prophylactic bracing decreases ankle injuries in collegiate female volleyball players. *American Journal of Sports Medicine, 36*, 324-327. doi: 10.1177/0363546507308358

Perrin, D. H. (2012). *Athletic taping and bracing* (3rd ed.). Champaign, Il: Human Kinetics

Petersen, W., Rembitzki, I. V., Koppenburg, A. G., Ellermann, A., Liebau, C., Brüggemann. G. P., & Best, R. (2013). Treatment of acute ligament injuries: A systematic review. *Arthroscopy and Sports Medicine, 133*, 1129-1141. doi: 11007/s00402-013-1742-5

Pienkowski, D., McMorrow, M., Shapiro, R., Caborn, D. M. N., & Stayton, J. (1995). The effect of ankle stabilizers on athletic performance: a randomized prospective study. *American Journal of Sports Medicine, 23*, 757-762.

Powers, S. K., & Howley, E. T. (2012). *Exercise physiology: Theory and application to fitness and performance* (8th ed.). New York, New York: McGraw-Hill

Prilutsky, B. I., & Zatsiorsky, V. M. (1994). Tendon action of two joint muscles: transfer of mechanical energy between joints during jumping, landing and running. *Journal of Biomechanics, 27*(1), 25-34.

Procter, P., & Paul, J. (1982). Ankle joint biomechanics. *Journal of Biomechanics, 15*, 627-634.

Rasmussen, O., & Tovborg-Jensen, I. (1982). Mobility of the ankle joint: Recording of rotary movements in the talocrural joint in vitro with and without the lateral collateral ligaments of the ankle. *Acta Orthopaedica Scandinavica, 53*, 155-160. http://dx.doi.org/10.3109/17453678208992194

Reid, D. C. (1992). *Sports injury and rehabilitation*. New York, NY: Churchill and Livingston.

Robbins, S., & Waked, E. (1998). Factors associated with ankle injuries: Preventative measures. *Sports Medicine, 25*(1), 63-72.

Roos, K. G., Kerr, Z. Y., & Mauntel, T. C. (2016). The epidemiology of lateral ligament complex ankle sprains in national collegiate athletic association sports. *American Journal of Sports Medicine, 45*, 201-208. http://dx.doi.org/11177%2F0363546516660980

Root, M. L., Orien, W. P., & Weed, J. H. (1977). *Normal and abnormal function of the foot*. Los Angeles: Clinical Biomechanics Corporation.

Rosenbaum, D., Kamps, N., Bosch, K., Thorwesten, L., Völker, K., & Eils, E. (2005). The influence of external ankle braces on subjective and objective parameters of performance in a sports-related agility course. *Knee Surgery Sports Traumatology Arthroscopy, 13*, 419-425. doi: 10.1007/s00167-004-0584-7 ·

Safran, M. R., Benedetti, R. S., Bartolozzi, A. R. 3rd., & Mandelbaum, B. R. (1999). Lateral ankle sprains: A comprehensive review: Part 1: Etiology, pathoanatomy, histopathogenesis, and diagnosis. *Medicine and Science in Sports and Exercise, 31*, 429-437.

Smith, B., Claiborne, T., & Liberi, V. (2016). Ankle bracing decreases vertical jump height and alters lower extremity kinematics. *International Journal of Athletic Therapy and Training, 21*(2), 39-46. https://doi.org/11123/ijatt.2014-0143

Santilli, V., Franscarelli, M. A., Paoloni, M., Franscarelli, F., Cameroto, F., De Natale, L., & De Santos, F. (2005). Peroneus longus muscle activation pattern during gait cycle in athletes affected by functional ankle instability: A surface electromyographic study. *American Journal of Sports Medicine, 33*, 1183-1187. doi: 10.1177/0363546504274147

Seegmiller, J. G., & McCaw, S. T. (2003). Ground reaction forces among gymnasts and recreation athletes in drop landings. *Journal of Athletic Training, 38*, 311-314.

SENIAM. (2014). *Surface electromyography for the non-invasive assessment of muscles*. Retrieved from http://seniam.org/

Shah, S., Thomas, A.C., Noone, J. M., Blanchette, C. M., & Wikstrom, E. A. (2016). Incidence and cost of ankle sprains in United States emergency departments. *Primary Care, 8*, 547-552. https://doi.org/10.1177/1941738116659639

Shima, N., Maeda, A., & Hirohashi, K. (2005). Delayed latency of peroneal reflex to sudden inversion with ankle taping or bracing. *International Journal of Sports Medicine, 26*, 476-480. doi: 10.1055/s-2004-821064

Shrout, P. E., & Fleiss, J. L. (1979). Intraclass correlations: uses in assessing rater reliability. *Psychological Bulletin, 86,* 420-428.

Simpson, K. J., Cravens, S., Higbie, E., Theodorou, C., & Delrey, P. (1999). A comparison of the Sport Stirrup, Malleoloc, and Swede-o® ankle orthoses for the foot-ankle kinematics of a rapid lateral movement. *International Journal of Sports Medicine, 20*, 396-402.

Simpson, K. J., Yom, J. P., Fu, Y-C., Arnett, S. W., O'Rourke, S., & Brown, C. N. (2013). Does wearing a prophylactic ankle brace during drop landings affect lower extremity kinematics and ground reaction forces*? Journal of Applied Biomechanics, 29*, 205-213. http://dx.doi.org/10.1123/jab.29.2.205

Sitler, M., Ryan, J., Wheeler, B., McBride, J., Arciero, R., Anderson, J., & Horodyski, M. (1994). The efficacy of a semirigid ankle stabilizer to reduce acute ankle injuries in basketball. *American Journal of Sports Medicine, 22*, 454-461. doi: 10.1177/036354659402200404

Slatyer, M. A., Hensley, M. J., & Lopert, R. (1997). A randomized controlled trial of piroxicam in the management of acute ankle sprain in Australian regular army recruits: The kapooka ankle sprain. *American Journal of Sports Medicine, 25*, 544-553. doi: 10.1177/036354659702500419

Smith, B., Claiborne, T., & Liberi, V. (2016). Ankle bracing decreases vertical jump height and alters lower extremity kinematics. *International Journal of Athletic Therapy and Training, 21*(2), 39-46. https://doi.org/11123/ijatt.2014-0143

Stasinopoulos, D. (2004). Comparison of three preventative methods in order to reduce the incidence of ankle inversion sprains among female volleyball players. *British Journal of Sports Medicine, 38*, 182-185. doi: 10.1136/bjsm.2002.003947

Stegeman, D. F., & Hermens, H. J. (1998). Standard for surface electromyography: the European Project (SENIAM). In: Hermens H.J., Rau G., Disselhorst-Klug C., Freriks B. (Eds.), *Surface Electromyography Application Areas and Parameters* (108-112). Proceedings of the Third General SENIAM Workshop on surface electromyography.

Stickley, C. D. (2017). The effects of ankle braces and taping on lower extremity running kinematics and energy expenditure in healthy, non-injured adults. *Gait and Posture, 58*, 108-114. https://doi.org/10.1016/j.gaitpost.2017.07.041

Surve, I., Schwellnus, M. P., Noakes, T., & Lombard, C. (1994). A fivefold reduction in the incidence of recurrent ankle sprains in soccer players using the sport-stirrup orthosis. *American Journal of Sports Medicine, 22*, 601-606.

Tamura, K., Radzak, K. N., Vogelpohl, R. E., Wisthoff, B. A., Oba, Y., Hetzler, R. K., & Stickley, C. D. (2017). The effects of ankle braces and taping on lower extremity running kinematics and energy expenditure in healthy, non-injured adults. *Gait and Posture, 58*, 108-114. DOI: https://doi.org/10.1016/j.gaitpost.2017.07.041

Tang, Y. M., Wu., Z. H., Liao, W. H., & Chan, K. M. (2010). A study of semi-rigid support on ankle supination sprain kinematics. *Scandinavian Journal of Medicine and Science in Sports, 20*, 822-826. doi: 10.1111/j.1600-0838.2009.00991.x

Thonnard, J. L., Bragard, D., Willems, P. A., & Plaghki, L. (1996). Stability of the braced ankle- A biomechanical investigation. *American Journal of Sports Medicine, 24*, 356-361. https://doi.org/10.1177/036354659602400318

Tiberio, D. (1987). The effect of excessive subtalar joint pronation on patellofemoral mechanics: A theoretical model. *Journal of Orthopaedic and Sports Physical Therapy.*

Tortora, G. J., & Nielson, M. (2010). *Principles of human anatomy* (4th ed.). Jefferson City, U.S.A: John Wiley and Sons, Inc.

Tropp, H., Odenrisk, P., & Gillquist, J. (1985). Stabilometry recordings in functional and mechanical instability of the ankle joint. *International Journal of Sports Medicine, 6*, 180-182. doi: 10.1055/s-2008-1025836

Urabe, Y., Kobayashi, R., Sumida, S., Tanaka., K., Yoshida, N., Nishiwaki, G. A., Tsutsumi, E., & Ochi, M. (2005). Electromyographic analysis of the knee during jump landing in male and female athletes. *The Knee, 12*, 129-134. https://doi.org/10.1016/j.knee.2004.05.002

Vanesky, K., Docherty, C. L., Dapena, J., & Schrader, J. (2006). Prophylactic ankle braces and knee varus-valgus and internal-external rotation torque. *Journal of Athletic Training*, *41*, 239-244.

Van Ingen Schenau, G. J., Bobbert, M. F., Huijing, P. A., & Woitiez, R. D. (1985). The instantaneous torque-angular velocity relation in plantar flexion during jumping. *Medicine and Science in Sports and Exercise, 17*, 422-426. *9*, 160-165.

Verbruggee, J. D. (1996). The effects of semirigid air-stirrup bracing vs. adhesive ankle taping on motor performance. *Journal of Orthopaedic & Sports Physical Activity, 23*(5), 320-325.

Verhagen, E. A. L. M., & Bay, K. (2010). Optimising ankle sprain prevention: a critical review and practical appraisal of the literature. *British Journal of Sports Medicine, 44*, 1082-1088. doi: 10.1136/bjsm.2010.076406

West, T., Ng, L., & Campbell, A. (2013). The effect of ankle bracing on knee kinetics and kinematics during volleyball specific tasks. *Scandinavian Journal of Medicine and Science in Sports, 24*, 958-963. doi:10.1136/bjsm.33.6.409

Wernli, K., Ng, L.,Phan, X., Davey, P., & Grisbrook, T. (2016). The relationship between landing sound, vertical ground reaction force, and kinematics of the lower limb during drop landings in healthy men. *Journal of Orthopaedic and Sports Physical Therapy, 46*, 194-199. doi: 10.2519/jospt.2016.6041

Wiesler, E. R., Hunter, D.M, Martin, D. F., Curl, W. W., & Hoen, H. (1996). Ankle flexibility and injury patterns in dancers. *American Journal of Sports Medicine, 24*, 754-757. doi: 10.1177/036354659602400609

Wiley, J. P., & Nigg, B. M. (1996). The effect of an ankle orthosis on ankle range of motion and performance. *Journal of Orthopaedic & Sports Physical Therapy, 23*, 362-369.

Wilson, D. H. (1972). Treatment of soft-tissue injuries by pulsed electrical energy. *British Medical Journal*, *2*, 269-270.

Wulf, G., Dufek, J. S., Lozano, L., Pettigrew, C. (2010). Increased jump height and reduced emg activity with an external focus. *Human Movement Science, 29*, 440-448. https://doi.org/10.1016/j.humov.2009.11.008

Yang, J., Marshall, S. W., Bowling, J. M., Runyan, C. W., Mueller, F. O., & Lewis, M. A. (2005). Use of discretionary protective equipment and rate of lower extremity injury in high school athletes. *American Journal of Epidemiology, 161*, 511-519. doi:10.1093/aje/kwi077

Yeow, C. H., Ng, Y. H., Lee., P. V., & Goh, J. C. (2010). Tibial cartilage damage and deformation at peak displacement compression during simulated landing impact. *American Journal of Sports Medicine, 38*, 816-823. doi:10.1177/0363546509350465

Yu, B., Lin, C-F., & Garret, W. E. (2006). Lower extremity biomechanics during the landing of a stop-jump task. *Clinical Biomechanics, 21*, 297-305. http://doi.org/10.1016/j.clinbiomech.2005.11.003

Zehr, E. P., Nakajima, T., Barss, T., Klarner, T., Miklosovic, S., Mezzarane, R. A., Nurse, M., & Komiyama, T. (2014). Cutaneous stimulation of discrete regions of the sole during locomotion produces "sensory steering" of the foot. *BMC Sports Science, Medicine, and Rehabilitation. 6*(33), https://doi.org/10.1186/2052-1847-6-33

Zhang, S. N., Bates, B. T., & Dufek, J. S. (2000). Contribution of lower extremity joints to energy dissipation during landings. *Medicine and Science in Sports and Exercise, 32*, 812-819. doi:10.1097/00005768-200004000-00014

Appendices

Appendix A

Recruitment Poster

Lakehead UNIVERSITY | School of Kinesiology

PARTICIPANTS WANTED FOR STUDY

Ankle Bracing's Effect on Lower Extremity Biomechanics During Athletic Performance Measures:

An Electromyographic and Kinetic Analysis

Conducted by:

Zachariah Henderson
School of Kinesiology
Lakehead University

- **Healthy, active individuals between the ages of 18 and 30 who participate in at least 150 minutes of moderate physical activity per week**
- **Previous experience with jumping or running activities (i.e., volleyball, basketball, track and field)**
- **Have not been diagnosed with a lower body or low back injury (fracture, sprain, strain, surgery) in the past 6 months that prevents participation in jumping or running**

All measurements are non-invasive and all information collected is confidential and anonymous. You will be tested during two 60 minute sessions. At these sessions, you will be asked to perform several vertical jumps, as well as a running task where you will quickly change direction. Each of these tasks will be peformed over a force platform and with sensors placed on your leg. Your vertical jump height, time to complete the running task, muscle activity, and movement forces will be measured with and without the application of an ankle brace.

For more information, or to volunteer please contact Zachariah Henderson

zhender1@lakeheadu.ca

Supervisor: Dr. Paolo Sanzo
psanzo@lakeheadu.ca

This study has been approved by the Lakehead University Research Ethics Board

Appendix B

Information Letter

Lakehead UNIVERSITY

School of Kinesiology

t: (807) 343 8544
kinesiology@lakeheadu.ca

Dear Potential Participant,

Thank you for expressing an interest in the study titled "Ankle Bracing's Effect on Lower Extremity Biomechanics During Athletic Performance: An Electromyographic and Kinetic Analysis." I am a Lakehead University graduate student in the School of Kinesiology, and will be undertaking this study under the supervision of Dr. Paolo Sanzo, Associate Professor in the School of Kinesiology. You have been invited to participate in this study because you are a physically active individual, who participates in at least 150 minutes of physical activity per week, have previous experience with jumping or running activities (i.e., volleyball, basketball, track and field), and are between the ages of 18-30 years. You are also not suffering from a lower body or low back injury, such as a strain or sprain that prevents you from running or jumping. You must not have suffered an ankle sprain in the last six months. Additionally, you have not suffered a fracture or undergone a lower body or low back surgery in the past six months. You must also not be allergic to any of the materials commonly used in ankle braces (Velcro, plastic), adhesive materials, or currently be pregnant. These are the criteria for individuals to be included in this study.

The purpose of study will be to see if wearing ankle braces affects muscle activation and movement forces in 42 healthy, active individuals. This will be done by having you perform a jumping task (Vertical Jump Test), as well as a task that involves quickly changing direction on top of a force platform while running (cutting task). Additionally, sensors will be placed on your leg muscles to measure muscle activation during these tasks. You will perform these tasks with and without two different types of ankle braces.

As a participant, you will be asked to sign up for two testing sessions, at least 24 hours, but no more than one week apart. Both sessions will last approximately 60 minutes and will take place in room SB-1028 of the Sanders Building at Lakehead University. Upon arriving for your testing session, you will be asked to complete and sign a consent form indicating that you know your rights as a participant. At this time you will also have the opportunity to ask any questions that you may have about the study. You will then be asked to fill out a Physical Activity Readiness Questionnaire (PAR-Q) to make sure that you can safely participate in physical activity and this study. You will also be asked to fill out a short questionnaire about your ankle brace and physical activity background, as well as the Functional Ankle Ability Measure (FAAM), a self-reported ankle health questionnaire. Before beginning the active portion of the study, you will be assigned the order that you will wear the ankle braces in for the jumping and cutting tasks.

The active portion of the study begins with a five minute warmup on a stationary bike. After the warmup, the researcher will place small sensors on your leg muscles and lower back. Once the sensors have been applied, you will perform a series of muscle contractions while the researcher provides resistance. Following this, you will perform the Vertical Jump Test with your assigned first ankle brace condition (no brace, ASO EVO™, or Active Ankle™ T1) over a force platform six times; three warmup jumps, followed by three best effort jumps. After you have completed the Vertical Jump Test with the first ankle brace condition, you will put on the next ankle brace condition and perform six trials again. Following completion of the Vertical Jump Test with the second ankle brace condition, you will then move on to the last ankle brace condition and perform six trials of the Vertical Jump Test one more time. A total of 18 jumps will be completed in 30 minutes. The session will be completed after a five minute cooldown on a stationary bike.

After the completion of the first testing session, you will be asked to come back one more time, 24 hours to one week after the first testing session. The procedures for the second day will mirror the first day. Instead of a Vertical Jump Test, though, you will be asked to perform a cutting task as quickly as possible, involving forward, side-ways, and backward movements.

As a researcher, it is my responsibility to inform you of your rights as a participant. You must understand that as a participant, your participation is completely voluntary. You may refuse to answer any questions or refuse to participate in any activity at any time throughout the study. You may also withdraw from this study at any time without penalty. Each participant has the right to remain anonymous. To ensure this, all results from this study will be presented anonymously, and your name will be replaced with a unique number. The data will only be accessible to the researchers conducting this study (i.e., Dr. Paolo Sanzo and Zachariah Henderson). The data will be stored at Lakehead University for five years in a locked filing drawer in the office of the supervisor. If the data is presented to the public as a written report or publication, or in the form of a verbal presentation, all participant data will remain anonymous.

There will be no direct physical harm to you as a participant completing this study. However, as with all physical activity, there is a minor risk that you may sustain a sprain or strain as a result of participating in these tests. Furthermore, as 18 jumps will be performed in a short period of time, there is a minor risk that fatigue may contribute to a possible sprain or strain. This risk will be minimized as you will perform the activity on a safe, dry, flat, and even surface under the direct supervision of the graduate researcher. This will assist in preventing any falls or other physical harm. Additionally, five minutes of rest will be provided when changing ankle brace conditions to allow for physical recovery. As you are a healthy and active individual, any potential risks will be minimized. In the unlikely event that you do sustain an injury while participating in this study, you will be referred to Lakehead University Student Health and Counselling. Additionally, prior to being admitted into the study, you will be asked whether you are fit to exercise, and whether or not you have sustained a lower body or low back injury in the last six months. If you indicate that you are not fit to exercise, or that you have sustained a lower body or low back injury in the past six months that prevents you from running and jumping, then you will not be eligible to participate in the study.

The results from this study may directly benefit society in that they may reveal the effects of wearing ankle braces on athletic performance measures, muscle activation of the lower body, and physical forces. Furthermore, any potential effect that previous experience with ankle braces and measures may come to light. This information can potentially be used by athletes, coaches, and healthcare providers when considering the prophylactic use of ankle braces. If you do wear ankle braces, there is direct benefit to you as a participant in that you will know the effects of ankle bracing on your own performance, muscle activation of the lower body, and physical forces when completing a maximal vertical jump or agility task.

Participants will be made aware of the results prior to publication or public presentation, if requested. Please contact me via email (given below) if you wish to receive a copy of the results. If you wish to participate in the study or have any additional questions or concerns then please do not hesitate to contact me directly or via email. This study has been approved by the Lakehead University Research Ethics Board. If you have any questions related to the ethics of the research and would like to speak to someone outside of the research team, please contact Sue Wright at the Research Ethics Board at 807-343-8283 or research@lakeheadu.ca.

Thank you for your consideration,

Zachariah Henderson, HBK
zhender1@lakeheadu.ca
(905) 995-8350

Dr. Paolo Sanzo
psanzo@lakeheadu.ca
(807) 343-8647

Appendix C

Consent Form

School of Kinesiology
t: (807) 343-8544
kinesiology@lakeheadu.ca

I_______________________________________ have read and understand the terms and conditions of this research study as outlined in the information letter.

- I willingly agree to participate in this study.
- I understand the potential risks and benefits of the study.
- I understand that I have certain rights as a participant in this study.
- I understand that as a volunteer, I may withdraw at any time throughout the study and may refuse to perform any activities.
- I am aware that the data recorded in this study will be securely stored at Lakehead University for five years after the study is complete.
- I have been informed that the results from this study will be made available to me via email once the study has been completed.
- I understand that personal information used will remain anonymous and will only be used by the researchers conducting the study.
- I understand that I will be protected and remain anonymous if publication or public presentation of the research findings should occur.
- I give my permission for my anonymous data to be published and presented publicly.

_______________________________________ __________________

(Signature of Participant) (Date)

_______________________________________ __________________

(Signature of Researcher) (Date)

☐ I would like to receive a written summary of the results.

Email address: ____

Appendix D

Physical Activity Readiness Questionnaire (Par-Q)

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO	
☐	☐	**1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	**2. Do you feel pain in your chest when you do physical activity?**
☐	☐	**3. In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	**4. Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	**5. Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	**6. Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	**7. Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ____

SIGNATURE ____________________________________ DATE ____________________

SIGNATURE OF PARENT ____________________________ WITNESS ____________________
or GUARDIAN (for participants under the age of majority)

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

Appendix E

Functional Ankle Ability Measure (FAAM)

Foot and Ankle Ability Measure (FAAM)
Activities of Daily Living Subscale

Please Answer **every question** with **one response** that most closely describes your condition within the past week.
If the activity in question is limited by something other than your foot or ankle mark "Not Applicable" (N/A).

	No Difficulty	Slight Difficulty	Moderate Difficulty	Extreme Difficulty	Unable to do	N/A
Standing	☐	☐	☐	☐	☐	☐
Walking on even Ground	☐	☐	☐	☐	☐	☐
Walking on even ground without shoes	☐	☐	☐	☐	☐	☐
Walking up hills	☐	☐	☐	☐	☐	☐
Walking down hills	☐	☐	☐	☐	☐	☐
Going up stairs	☐	☐	☐	☐	☐	☐
Going down stairs	☐	☐	☐	☐	☐	☐
Walking on uneven ground	☐	☐	☐	☐	☐	☐
Stepping up and down curbs	☐	☐	☐	☐	☐	☐
Squatting	☐	☐	☐	☐	☐	☐
Coming up on your toes	☐	☐	☐	☐	☐	☐
Walking initially	☐	☐	☐	☐	☐	☐
Walking 5 minutes or less	☐	☐	☐	☐	☐	☐
Walking approximately 10 minutes	☐	☐	☐	☐	☐	☐
Walking 15 minutes or greater	☐	☐	☐	☐	☐	☐

Foot and Ankle Ability Measure (FAAM)
Activities of Daily Living Subscale
Page 2

Because of your foot and ankle how much difficulty do you have with:

	No Difficulty at all	Slight Difficulty	Moderate Difficulty	Extreme Difficulty	Unable to do	N/A
Home responsibilities	☐	☐	☐	☐	☐	☐
Activities of daily living	☐	☐	☐	☐	☐	☐
Personal care	☐	☐	☐	☐	☐	☐
Light to moderate work (standing, walking)	☐	☐	☐	☐	☐	☐
Heavy work (push/pulling, climbing, carrying)	☐	☐	☐	☐	☐	☐
Recreational activities	☐	☐	☐	☐	☐	☐

How would you rate your current level of function during you usual activities of daily living from 0 to 100 with 100 being your level of function prior to your foot or ankle problem and 0 being the inability to perform any of your usual daily activities.

__ __ __ . 0 %

Martin, R; Irrgang, J; Burdett, R; Conti, S; VanSwearingen, J: Evidence of Validity for the Foot and Ankle Ability Measure. Foot and Ankle International. Vol.26, No.11: 968-983, 2005.

Foot and Ankle Ability Measure (FAAM)
Sports Subscale

Because of your foot and ankle how much difficulty do you have with:

	No Difficulty at all	Slight Difficulty	Moderate Difficulty	Extreme Difficulty	Unable to do	N/A
Running	☐	☐	☐	☐	☐	☐
Jumping	☐	☐	☐	☐	☐	☐
Landing	☐	☐	☐	☐	☐	☐
Starting and stopping quickly	☐	☐	☐	☐	☐	☐
Cutting/lateral Movements	☐	☐	☐	☐	☐	☐
Ability to perform Activity with your Normal technique	☐	☐	☐	☐	☐	☐
Ability to participate In your desired sport As long as you like	☐	☐	☐	☐	☐	☐

How would you rate your current level of function during your sports related activities from 0 to 100 with 100 being your level of function prior to your foot or ankle problem and 0 being the inability to perform any of your usual daily activities?

__ __ __ . 0%

Overall, how would you rate your current level of function?

☐ Normal ☐ Nearly Normal ☐ Abnormal ☐ Severely Abnormal

Martin, R; Irrgang, J; Burdett, R; Conti, S; VanSwearingen, J: Evidence of Validity for the Foot and Ankle Ability Measure. Foot and Ankle International. Vol.26, No.11: 968-983, 2005.

Appendix F

Background Questionnaire

Lakehead UNIVERSITY

School of Kinesiology
t: (807) 343-8544
f: (807) 343-8944
kinesiology@lakeheadu.ca

Background Questionnaire

Participant I.D.	
Sex	
Age (years)	
Height (cm)	
Weight (kg)	
Dominant foot (L/R)	

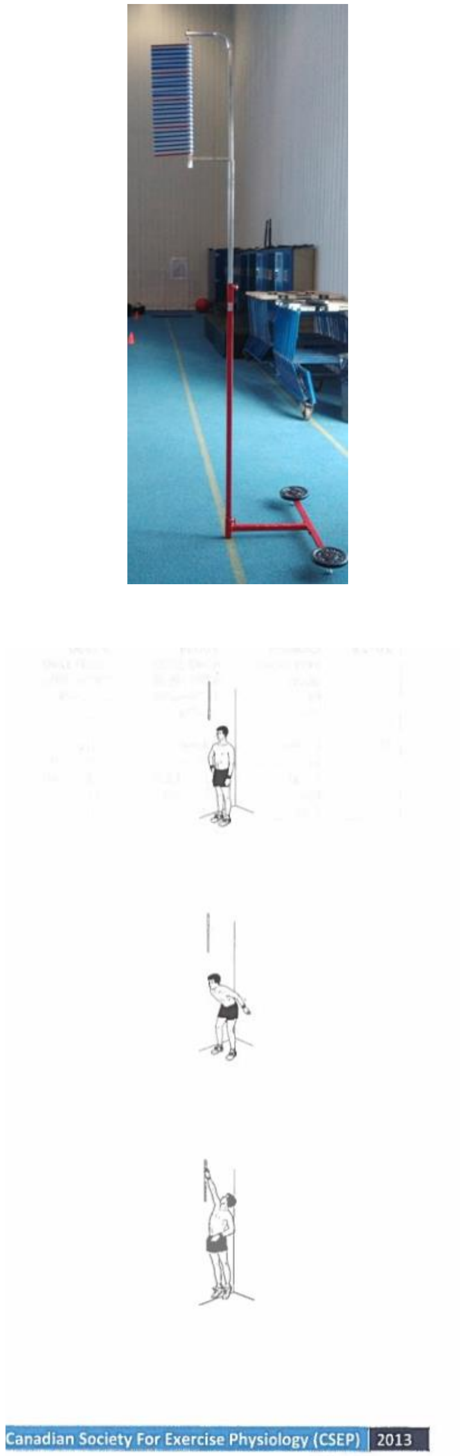
1) Have you ever taped your ankle(s) or used an ankle brace(s)? If so, what kind and for how long?

2) What sport or activity would you say you participate in the most?

3) Of the braces you used, what one did you prefer? Why?

Appendix G

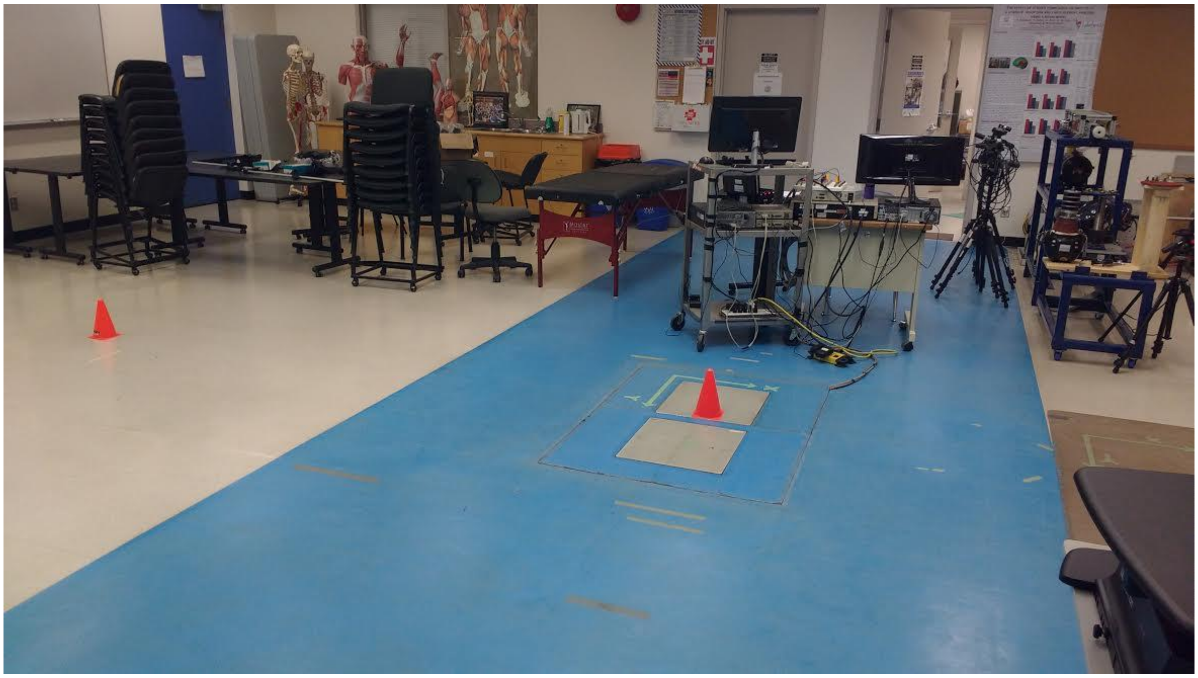
Vertical Jump Test Diagram and Set-up With Vertec ™ Device

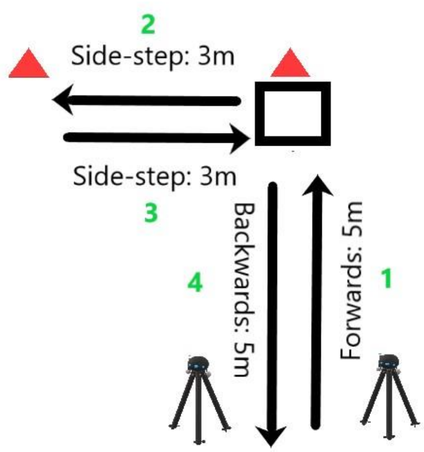


Vertical Jump Test. This figure displays the equipment and procedures for the Vertical Jump Test. Adapted from “Canadian society for exercise physiology- physical activity training for health” by CSEP, p. 57.

Appendix H

Cutting Task Diagram and Set-up




2
Side-step: 3m
Side-step: 3m
3
Backwards: 5m
4
Forwards: 5m
1

Appendix I

Borg Rating of Perceived Exertion

#	Level of Exertion
6	No exertion at all
7	
7.5	Extremely light (7.5)
8	
9	Very light
10	
11	Light
12	
13	Somewhat hard
14	
15	Hard (heavy)
16	
17	Very hard
18	
19	Extremely hard
20	Maximal exertion

BORG RPE. This figure visually represents the BORG rating of perceived exertion scale. Adapted from "Perceived Exertion (Borg Rating of Perceived Exertion Scale)" by The Centers for Disease Control and Prevention. Retrieved from https://www.cdc.gov/physicalactivity/basics/measuring/exertion.htm

Appendix J

Pilot Study

The Biomechanical Effects of Ankle Bracing During

Performance Measures— A Pilot Study

Zachariah Henderson

Dr. Paolo Sanzo

Lakehead University

KINE-5070

Abstract

Introduction: Ankle braces reduce ankle injuries by restricting range of motion (ROM) at the ankle. However, the proximal effects of restricting ankle ROM during functional activities remains unclear. It has been suggested that ankle brace use can increase the risk of lower extremity injury above the ankle, alter lower extremity kinematics, and increase ground reaction forces (GRFs). Recent research has also revealed reductions in vertical jump height when wearing ankle braces, indicating that muscular activation of the lower extremity may be affected by wearing ankle braces. Therefore, the purpose of this pilot study was to determine if the protocol could detect changes in lower extremity and lumbar spine electromyography (EMG) and GRFs during the Vertical Jump Test and a lateral cutting task.
Method: Ten physically active individuals (23.2 years, +/- 1.095 years) completed the Vertical Jump Test and a lateral cutting task under three conditions (no brace, ASO EVO ™ brace, and T1 brace). Electromyographic muscle activity of the peroneus longus (PL), lateral gastrocnemius (LG), biceps femoris (BF), rectus femoris (RF), gluteus medius (GM), erector spinae (ES), and vertical GRF was recorded for the Vertical Jump Test during take off and landing. Electromyography, vertical, anteroposterior, and mediolateral GRFs were recorded during directional changes for the lateral cutting task. The Friedman Test was used to examine the effect of the independent variable (brace condition) on the dependent variables (EMG, GRFs, vertical jump height, and time to complete the lateral cutting task) with an alpha level set at $p < .05$
Results: During the lateral cutting task, there was a significant decrease in normalized RF mean EMG activity during the contact phase of the first force platform contact (1a), X^2 (2) = 9.314, p = .009. Normalized RF mean EMG activity during 1a decreased when wearing both the AE (M = 259.4725, SD = 128.28) and T1 (M= 195.51 SD = 98.28) braces when compared to the no brace condition (M = 583.57, SD = 590.83). Post hoc analysis revealed a significant (p = .015) decrease in RF mean EMG activity during 1a when wearing the T1 ankle braces, with a large effect size (r = .5). There was no significant difference between conditions for any of the other variables.
Conclusions: The results of this study provide the basis for future studies examining lower extremity EMG activity, GRFs, and performance when wearing ankle braces during functional performance measures. Rectus femoris EMG activity significantly decreased when wearing a brace during the lateral cutting task, indicating that ankle braces may indeed affect EMG activity of muscles above the ankle. Future studies should utilize a larger sample size and recruit participants who have experience wearing ankle braces to confirm these results.

Introduction

Sprains of the lateral ankle ligaments represent the most frequently reported injury in the National Collegiate Athletics Association (NCAA; Roos, Kerr, & Mauntel, 2016). Currently, the National Athletic Trainers Association recommends that ankle braces be worn by athletes returning to play from an ankle sprain (Kaminski et al., 2013). Recent systematic reviews by Leppänen, Aaltonen, Parkkari, Heinonen, and Kujala, (2014) and Petersen et al. (2013) of ankle bracing and ankle injury have concluded that, in athletic populations, both softshell and semi-rigid ankle braces can effectively prevent and treat ankle injuries. In a study of ankle brace use and ankle injury prevention in basketball players, however, McGuine et al. (2011) noted an 85% increase in non-ankle injuries of the lower extremity when wearing softshell ankle braces. Furthermore, recent research has revealed decreases in vertical jump height when wearing softshell and semi-rigid ankle braces (Henderson, Sanzo, & Zerpa, 2016; Parsley, Chinn, Lee, Ingersoll, & Hertel, 2013; Smith, Claiborne, & Liberi, 2016). Decreases in agility performance when wearing softshell and semi-rigid ankle braces have also been observed by Ambegaonkar et al. (2011). As such, it has been suggested that further research be done on the effects of ankle braces on lower extremity and lumbar spine biomechanics (McGuine et al., 2011), as well as performance measures (Ambegaonkar et al., 2011).

The primary cause of lateral ankle sprains is over-stretching of the lateral ankle ligaments (Ferran & Maffulli, 2006). As such, restricting the range of motion (ROM) of the ankle joint has long been considered the main mechanism by which ankle braces prevent ankle sprains (Verhagen & Bay, 2010). Ankle braces function in different ways to provide ROM restriction at the ankle, depending on the style of ankle brace. Softshell style ankle braces have a lace-up design, and may, or may not, feature heel-lock and horseshoe straps to restrict frontal and sagittal plane motions (Gudibanda & Wang, 2005). Conversely, semi-rigid style ankle braces allow for free motion in the sagittal plane via a hinge that sits underneath the heel. Unlike the softshell ankle brace, a plastic shell is present on the medial and lateral aspect of a semi-rigid ankle brace to restrict eversion and inversion (West, Ng, & Campbell, 2013). In physically active individuals, softshell ankle braces have been shown to effectively restrict plantarflexion and dorsiflexion during jump landing (DiStefano, Padua, Brown, & Guskiewicz, 2008), vertical jump take off (Smith et al., 2016), and agility like cutting tasks (Gudibanda & Wang, 2005). Semi-rigid ankle bracing has also demonstrated the ability to significantly restrict ankle inversion in competitive female basketball players during agility like cutting tasks (Klem, Wild, Williams, & Ng, 2016). While both semi-rigid and softshell ankle braces have been shown to restrict ankle ROM during functional tasks, the kinetic and proximal effects of restricting ankle ROM during jumping and agility like cutting tasks are not well understood.

The kinetic variable ground reaction forces (GRFs) represents the force exerted by the ground on a body (Nilsson & Thorstensson, 1989). It has been theorized that if GRFs exceed that which can be dissipated by the musculoskeletal system, injury to the lower extremity may occur (Dufek & Bates, 1990). Larger knee flexion angles (Dufek & Bates, 1990; Wernli, Ng, Phan, Davey, & Grisbrook, 2016), lower extremity extensor and flexor, eccentric and concentric contraction (Devita & Skelly, 1992), and increased dorsiflexion and plantarflexion ROM (Gross & Nelson, 1988) have been attributed to reduced GRFs during jump landings. During the squat, artificial reduction of plantarflexion ROM during the eccentric portion has been shown to result in decreased quadriceps muscle activity and increased soleus muscle activity (Macrum, Bell, Boling, Lewek, & Padua, 2012). As ankle braces are known to restrict dorsiflexion and plantarflexion, a similar reduction in quadriceps muscle activation may be present when wearing ankle braces during other functional tasks. Thus, the ability to attenuate GRFs when wearing ankle braces may be reduced during jump landings and other functional tasks, in addition to potential performance decreases. Despite this, no studies have examined the effects of ankle braces on muscular activation above the ankle during jumping, lateral cutting, or agility tasks. Furthermore, a limited number of studies have investigated the effects of ankle braces on kinetics and lower extremity kinematics during jumping, lateral cutting, and agility tasks.

In kinetic and kinematic studies of wearing ankle braces during functional tasks, results have been mixed. Hodgson, Tis, Cobb, and Higbie (2005) examined the effect of semi-rigid ankle braces on lower extremity kinematics and kinetics during a 0.61 metre drop landing in female collegiate volleyball players. No differences were found in knee and hip kinematics when completing the task with or without Active Ankle T2™ semi-rigid ankle braces; however, vertical GRF was significantly greater in the Active Ankle T2™ condition. In comparison, DiStefano et al. (2008) explored the kinematic and kinetic effects of wearing softshell ankle braces in recreational basketball and volleyball players during a 30 centimetre (cm) high broad jump landing. When broad jump distance was standardized to half of the participant’s height, knee flexion angle increased by 3 degrees at ground contact when wearing the AE softshell ankle braces. Ground reaction forces were not affected by wearing the AE ankle braces. Although these studies used different methodologies, it appears that wearing softshell and semi-rigid ankle braces may produce different kinematic and kinetic changes during jump landings. These differences are also present in kinematic and kinetic studies of agility and cutting when wearing softshell and semi-rigid ankle braces.

Klem et al. (2016) investigated the kinetic and kinematic effects of semi-rigid and softshell ankle braces in competitive female basketball players when performing a 90 degree cutting maneuver. Anterior shear forces at the knee were significantly increased during the deceleration phase of cutting when wearing the AE ankle brace, compared to the no brace control condition. Alternatively, when wearing the Active Ankle T2™ ankle brace, anterior shear forces were significantly reduced. Furthermore, knee internal rotation was significantly increased when wearing both the AE and Active Ankle T2™ ankle braces. These results contradict that of West et al. (2013) who also included a 90 degree cutting task in their kinetic and kinematic evaluation of semi-rigid ankle braces. In competitive female volleyball players, knee kinematics and GRFs were not significantly altered when wearing the Active Ankle™ T2 ankle brace. Similar to Klem et al. (2016), there was a decrease in lateral shear force at the knee when wearing Active Ankle T2™ ankle braces. Thus, it appears that a semi-rigid ankle brace may be effective in reducing shear forces at the knee during a 90 degree cutting maneuver. The effects on lower extremity kinematics, however, remains unclear.

When performing a vertical jump, several kinematic and kinetic variables have been associated with improved vertical jump height including: segmental angular displacement of the upper leg and trunk (Hsieh & Cheng, 2016), production of mechanical power (Aragón-Vargas & Gross, 1997), ankle angle at take-off (Aragón-Vargas & Gross, 1997), and minimum angle of the hip, knee, and ankle joint (Hsieh & Cheng, 2016). Therefore, potential alteration of lower extremity kinematics and kinetics by an ankle brace could impact vertical jump height by affecting these variables. Smith et al. (2016) investigated the effects of a McDavid 195T softshell ankle brace on Vertec™ measured vertical jump height, lower extremity kinematics, and EMG activity of the soleus and gastrocnemius muscles. In 20 varsity athletes, vertical jump height was reduced by .92 inches (in) when wearing the McDavid 195T ankle brace, representing a significant decrease. A reduction in ankle ROM and soleus muscle activity was also observed. The authors suggested that the reduction in soleus muscle activity was the result of decreased ankle ROM, which contributed to the reduction in vertical jump height. To the author's knowledge, this is the only study examining the effect of ankle bracing on performance measures that also included the measurement of EMG activity; however, the reduction in vertical jump height is consistently resonated in recent research.

Using a Vertec™ device, Henderson et al. (2016) noted a significant 2.95 cm decrease in vertical jump height when jumping sport athletes wore T1 semi-rigid ankle braces. A similar, though insignificant reduction in jump height was also observed when wearing AE ankle braces. No differences were observed between the no brace control, AE, or T1 conditions with respect to agility time, as measured by the T-test Agility Test. Parsley et al. (2013) had similar results examining functional performance when wearing a prototype external ankle brace, AE ankle braces, and Aircast Airsport™ semi-rigid ankle braces. In physically active males, vertical jump height was significantly reduced by 1.3-1.8 cm in all brace conditions, compared to the no brace control condition. Agility, as measured by the SEMO agility run, was not significantly affected. From these studies, it appears that vertical jump height is impaired, while agility time is not affected when wearing any type of ankle brace.The research using different ankle braces and methodologies presents conflicting evidence to this conclusion.

Ambegaonkar et al. (2011) investigated the effects of ankle tape, the Swedo-O Ankle Lok™ softshell ankle bracse, and Aircast Air-Stirrup™ semi-rigid ankle braces on vertical jump height and agility time. In healthy participants, vertical jump height was not affected as measured by the Sargent Jump Test. When wearing the Aircast Air-Stirrup™ ankle braces, however, agility time was increased by an average of .59 seconds during the Right-Boomerang Run Test. These results suggested that agility performance is negatively affected by wearing a semi-rigid ankle brace, whereas vertical jump performance remained unaffected; the opposite findings of Henderson et al. (2016) and Parsley et al. (2013). Furthering this discrepancy, Leonard and Rotay (2014) examined the effects of ankle tape, and generic softshell ankle braces on vertical jump height, power, and agility. As measured with a Vertec™ device, there was no difference in vertical jump height across conditions in athletic and non-athletic individuals. No significant differences were observed between conditions for the vertical jump power test, or for the Illinois Agility Test. As Ambegaonkar et al. and Leonard and Rotay used different ankle brace models compared to other studies, it appeared that the type and model of brace may affect results, furthering the need for research in this area.

Currently, no studies have examined GRFs and EMG activity of the lower extremity and lumbar spine musculature during performance measures, such as vertical jump and lateral cutting when wearing softshell and semi-rigid ankle braces. Given observed changes in the lower extremity kinematics, kinetics, and performance in some studies, EMG of the lower extremity and lumbar spine may be similarly affected when wearing ankle braces. Furthermore, it appears that the style of ankle brace may affect these variables. As such, the purpose of this pilot study was to determine if the protocol and tasks were feasible to detect changes in the biomechanics of the lower extremity when wearing two different types of ankle braces. Specifically, lower extremity and lumbar spine EMG muscular activity, GRFs, and performance during the Vertical Jump Test and a lateral cutting task.

Methods

Subject Inclusion/Exclusion Criteria

Ten participants were recruited for this study. See Table 1 for demographic and anthropometric information of participants.

Table 1. Participant demographic and anthropometric information

Gender	5 male, 5 female
Age (years)	M = 23.2, SD = 1.095
Height (cm)	M = 175.8, SD = 7.21
Body Mass (kg)	M = 74.4, SD = 7.58
Foot Dominance (left/right)	3 left, 7 right

After ethical approval was granted by the academic institution, potential participants were recruited via purposive and convenience sampling. Prospective participants were included into this study if they 1) were male or female, recreationally active students at Lakehead University; 2) participated in at least 150 minutes of moderate to vigorous aerobic activity each week; and 3) were between the ages of 18-30 years. This specific population was selected to represent the physically active population at Lakehead University and to maximize recruitment potential. Participants were excluded from this study if they: 1) had suffered from a diagnosed ankle injury over the last six months (e.g., sprain, fracture, tendonitis); 2) were currently suffering from an acute and/or chronic lower extremity or low back injury (i.e., strain, sprain, herniated disk) that precluded them from participating in jumping or running activities; 3) had undergone any lower extremity or low back surgical procedure in the last six months; 4) were allergic or sensitive to adhesive tape or any of the material present in the AE and T1 ankle braces (i.e., Velcro, plastic); and 5) were pregnant.

Potential participants who expressed interest in participating in the study were provided with an electronic and hardcopy of the participant recruitment letter, outlining details of the study. Once the potential participant was deemed eligible to participate, testing sessions were scheduled at times mutually agreeable to the participant and the researcher.

Procedures

Data were collected during two testing sessions, approximately 24 hours apart. Participants performed the Vertical Jump Test during the first testing session and the lateral cutting task during the second testing session. The Vertical Jump Test and lateral cutting task were performed under three conditions: A no brace control, wearing the AE ankle braces, and wearing the T1 ankle braces on both ankles. Testing took place in the School of Kinesiology

Multipurpose Lab at Lakehead University. Participants were asked to wear appropriate clothing and footwear for physical activity. At the first testing session, the researcher provided a verbal overview of the study and answered any questions that the participant had. Once verbal consent was obtained, the participant signed the consent form and filled out a Physical Activity Readiness Questionnaire (PAR-Q) to determine if he/she was fit to exercise. Following written consent and the completion of the PAR-Q, demographic data were recorded including: age (years), height (cm), weight (kg), and foot dominance.

After obtaining written informed consent and recording the demographic information, participants completed a five minute warmup on a cycle ergometer, at an intensity of 10-12 on the Borg Scale of Perceived Exertion (CSEP, 2013). After the warmup, the wireless electrodes of a Delsys Trigno^{TM} Wireless EMG system were applied to the skin of the dominant leg of the participant, defined as the leg that the participant would kick a ball with. Electrodes were applied in order to collect EMG data from the: peroneus longus (PL), lateral gastrocnemius (LG), biceps femoris (BF), gluteus medius (GM), erector spinae (ES), and rectus femoris (RF) muscles. Electrode location was based on the Surface Electromyography for the Non-Invasive Assessment of Muscle (SENIAM; 2016) guidelines (Figure 1). Before applying the electrodes, the skin underlying the electrode sites were prepared by shaving and cleaning the area with isopropyl alcohol to help improve signal attenuation (Delsys Inc., 2012). Standard adhesive interfaces were used to attach the electrode to the participant's skin. Following electrode application, participants completed maximal voluntary contractions (MVC) for all corresponding muscles. To perform each MVC, the researcher provided manual resistance while the participant contracted the specified muscle for three seconds.

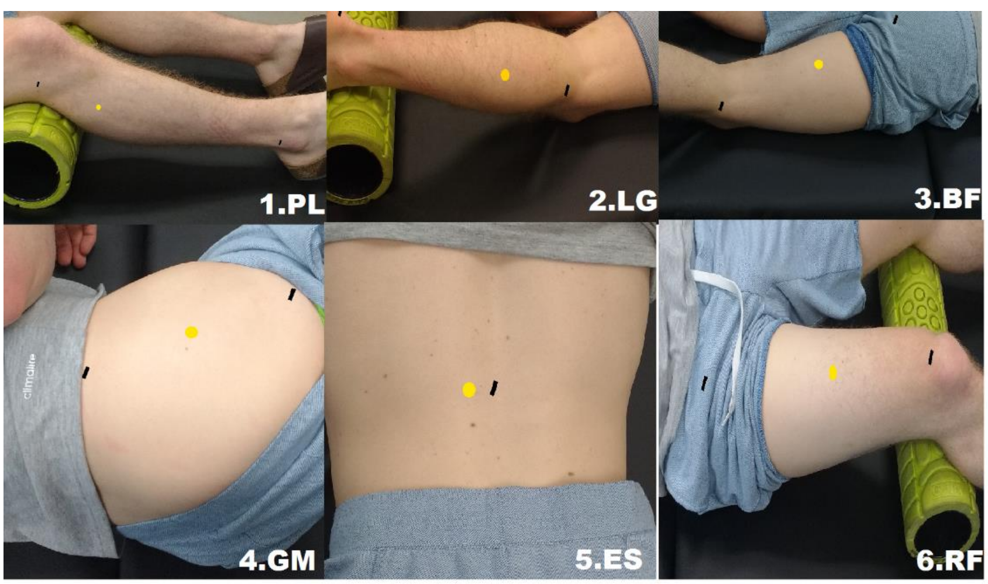


Figure 1. Electrode location. This figure displays the location for all 6 electrodes. The black lines represent anatomical landmarks. Yellow dots represent electrode locations.

After the MVCs were recorded, the researcher familiarized the participant with the Vertical Jump Test. The Vertical Jump Test was based on the CSEP (2013) guidelines and used to assess vertical jump height. Before beginning the Vertical Jump Test, the participant's standing reach height was recorded; participants stood erect over an AMTI force platform, perpendicular to a Vertec™ device. The participant then raised his/her arm overhead, touching as high as possible on the Vertec™ device with his/her arm that corresponded with the dominant leg. This height (in) was recorded, and then converted to cm by the researcher. After standing reach was recorded, the participant moved into position to perform his/her practice and recorded jumps.

To perform the Vertical Jump Test, the participant positioned his/her feet approximately shoulder width apart, with the dominant foot on the force platform. The participant initiated the jump by bending at the hip and knees and lowering into a 45 degree semi-squat position. While descending, the participant moved his/her arms in a counterweight fashion. In this position, the participant's form was visually evaluated by the researcher; if the participant's position was not acceptable (45 degrees), the researcher would stop the test. The participant held the 45 degree semi-squat position for two seconds, before jumping and touching the Vertec™ device as high as possible. Participants were required to land evenly on both feet, with the dominant foot on the force platform. If the participant landed off balance, the trial was not recorded.

Submaximal attempts were practiced for participants to become comfortable with the test procedure and allow the researcher to provide feedback regarding form. Once a participant was familiar with the test, he/she performed the first of three recorded trials. The researcher began collecting EMG and GRF data and notified the participant; after standing still for 3 seconds, the participant performed a maximal effort jump. If all key form components of the jump were met, vertical jump height (in), EMG activity, and GRF data were recorded. The participant then performed the test two more times, each spaced approximately one minute apart to allow for physical recovery.

After completion of the Vertical Jump Test with no ankle braces, participants had a five minute rest period. During this time, the participant applied an AE ankle brace to both ankles. Ankle braces were applied and appropriately sized based on the manufacturer's guidelines. Participants were allowed to adjust the ankle braces during the testing session, if required, to maintain the manufacturer's described fit. Ankle braces were applied without modifying the participant's normal training shoe.

Following the application of the AE ankle braces, the Vertical Jump Test was performed in the same manner as the control condition. After completion of the Vertical Jump Test with the AE ankle braces, another five-minute rest period was used to allow the participant to remove the AE ankle braces, apply T1 ankle braces, and physically recover from the task. Application of the T1 ankle braces followed the same procedures as the AE ankle braces. After application of the T1 ankle braces, the Vertical Jump Test was performed in the same manner as the previous two conditions. The testing session was concluded after completion of the Vertical Jump Test with the T1 ankle braces.

The second testing session occurred approximately 24 hours after the first testing session. Testing began in the same manner as the first session. Following the warmup and application of the electrodes, the lateral cutting task was explained to the participant (Figure 3). To perform the lateral cutting task, the participant positioned themselves at the start line, five metres away from the force platform. Here, Brower™ timing gates were set up, which were used to electronically record lateral cutting task time. As with the Vertical Jump Test, participants completed submaximal practice attempts to familiarize his/her self with the lateral cutting task procedure. To begin the lateral cutting task, the participant positioned themselves in an athletic stance position, with his/her dominant limb forward. The researcher began collecting EMG and GRF data and notified the participant; when the participant was ready, he/she initiated the test by crossing the timing gates. The participant ran forwards toward the force platform; a cone was positioned directly in front of the force platform for the participant to touch. The participant touched the cone, simultaneously planting the dominant foot on the force platform. Immediately after touching the cone, the participant side shuffled to his/her left or right, based on the planting foot, for three metres towards another cone. The participant touched this cone and then side shuffled back to the first cone. After touching the first cone and landing on the force plate for the second time, the participant then backpedaled towards the starting line. Participants cut the same direction for all trials. The test concluded once the participant crossed the timing gate at the starting line.

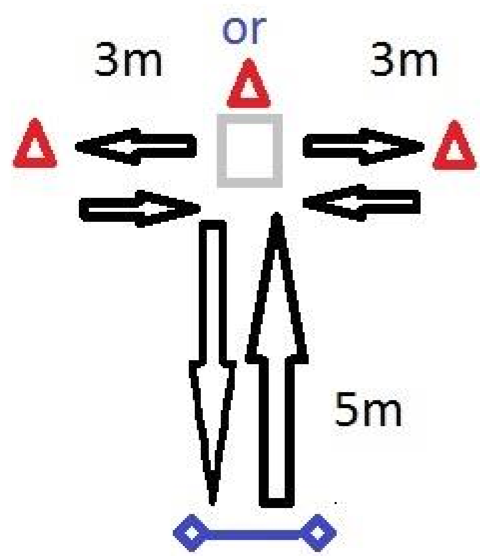


Figure 3. Lateral cutting task. This figure displays the set up for the lateral cutting task.

Ground reaction forces, EMG activity, and time to complete the lateral cutting task was recorded if the participant successfully touched the force platform twice and completed the test in its entirety. After the completion of three recorded trials with no brace, the testing session mirrored that of the first testing session; participants completed the lateral cutting task when wearing the AE and T1 ankle braces. The testing session was completed after the lateral cutting task had been performed under all three conditions.

Data Processing

Force platform and EMG data were collected simultaneously using PowerLab (16/30) hardware, Delsys Trigno™ Wireless EMG system, Delsys EMG Works™ software, and LabChart™ data acquisition software. An interface board was used to time synchronize force platform and EMG data. LabChart software was used to analyze EMG and GRF data. Raw EMG was rectified and filtered using a lowpass filter (12 Hz). Mean EMG activity for each muscle, measured in volts (V), was calculated and expressed as a percentage of MVC. Maximum GRFs, measured in newtons (N), were calculated and expressed as a percentage of bodyweight.

Vertical jump. For the purposes of EMG and GRF analysis, the Vertical Jump Test was divided into two phases; take off and landing. The take off phase was defined as the time at which vertical GRF began to increase (greater than 5 N) from the stationary system weight, to the time that system weight equaled 0 (+/- 5 N). The landing phase was defined as the time at which system weight increased from zero (greater than 5N) to the time that system weight returned to stationary level (+/- 5 N; see Figure 4).

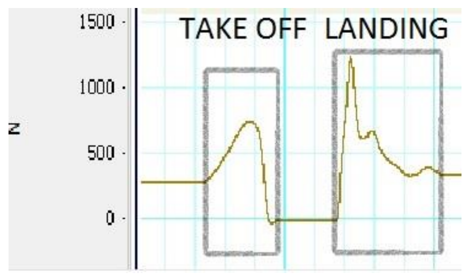


Figure 4. Vertical jump phases. This figure illustrates the take-off and landing phase of the Vertical Jump Test, based on vertical GRF.

Lateral cutting task. For the purposes of EMG and GRF analysis, the lateral cutting task was divided into two phases, based on two different force platform contacts. The first contact with the force platform involved deceleration in the sagittal plane, followed by lateral movement in the frontal plane. The deceleration phase (1a) of the first platform contact was defined as the time at which anteroposterior GRF began to increase (greater than 5 N) from 0 N to the time at which maximum anteroposterior GRF was reached (Houck, 2003). The propulsive phase (1b) was defined as the time between maximum anteroposterior GRF and a reading of 0 N (+/- 5 N) (Figure 5). The second force platform contact involved decelerating in the frontal plane, followed by backward movement in the sagittal plane. The deceleration phase (2a) for the second force platform contact was defined as the time at which mediolateral GRF began to increase (greater than 5 N) from 0 N to the time at which maximum mediolateral GRF was reached. The propulsive phase (2b) was defined as the time between maximum mediolateral GRF and a reading of 0 N (+/- 5 N; see Figure 6).

Figure 5. Lateral cutting task first contact anteroposterior GRF. This figure illustrates the deceleration phase (1a) and propulsive phase (1b) of the first contact with the force platform, based on anteroposterior GRF.

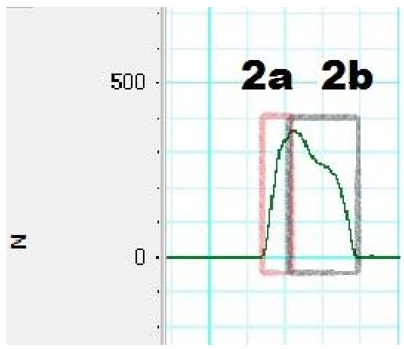


Figure 6. Lateral cutting task second Contact Mediolateral GRF. This figure illustrates the deceleration phase (2a) and propulsive phase (2b) of the second contact with the force platform, based on mediolateral GRF.

Statistical Analysis

Statistical analysis was performed using Statistical Package for the Social Sciences (SPSS) 24. Dependant variable values were averaged from the three trials for each participant. The Friedman Test was conducted to see if the independent variable (no brace, AE, and T1) affected each dependant variable. Alpha level was set at $p < .05$. Post hoc analysis was performed using a Wilcoxon signed-rank test. A Bonferroni correction was applied to each pairwise comparison, resulting in a significance level of $p = .017$. Effect size was calculated by dividing the Z value by the square root of the number of observations (n = 30). One participant's lateral cutting task RF EMG data were excluded from the analysis due to equipment malfunction.

Results

Descriptive Statistics

Descriptive statistics for performance measures are listed in Table 2. Descriptive statistics for EMG and GRF variables during the Vertical Jump Test are listed in Tables 3 and 4. Descriptive statistics for EMG and GRF variables during the lateral cutting task are listed in Tables 5-8.

Table 2

Performance Measures Descriptive Statistics

BRACE TYPE	NO BRACE		ASO EVO		AAT1	
	MEAN	SD	MEAN	SD	MEAN	SD
VERTICAL JUMP HEIGHT (cm)	265.19	17.19	264.51	16.19	263.32	16.75
CUTTING TASK TIME (sec)	6.49	0.69	6.38	0.72	6.36	0.75

*Significant difference between conditions (*p*<.05).

Table 3

Vertical Jump Take Off Descriptive Statistics

BRACE TYPE	NO BRACE		ASO EVO		AAT1	
	MEAN	SD	MEAN	SD	MEAN	SD
NORMALIZED PERONEUS LONGUS MEAN EMG	178.93	283.34	172.47	119.27	173.27	115.16
NORMALIZED L. GASTROCNEMIUS MEAN EMG	209.86	102.81	191.76*	92.26	201.62	94.90
NORMALIZED BICEPS FEMORIS MEAN EMG	53.66	23.76	53.87	21.76	73.38	92.18
NORMALIZED GLUTEUS MEDIUS MEAN EMG	161.32	98.65	142.97	69.96	134.97	52.65
NORMALIZED ERECTOR SPINAE MEAN EMG	256.10	283.34	163.29	119.27	180.09	115.16
NORMALIZED RECTUS FEMORIS MEAN EMG	458.53	220.10	414.08	173.95	410.77	185.23
NORMALIZED MAXIMUM VERTICAL GRF	112.46	9.17	111.22	9.81	111.44	11.38

All EMG values are expressed as a percentage of MVC. All GRF values are expressed as a percentage of bodyweight. *Significant difference between conditions (*p*<.05).

Table 4

Vertical Jump Landing Descriptive Statistics

BRACE TYPE	NO BRACE		ASO		AAT1	
	MEAN	SD	MEAN	SD	MEAN	SD
NORMALIZED PERONEUS LONGUS MEAN EMG	223.00	185.56	165.99	80.98	195.25	147.66
NORMALIZED L. GASTROCNEMIUS MEAN EMG	167.54	79.86	195.57	122.32	171.39	92.75
NORMALIZED BICEPS FEMORIS MEAN EMG	54.30	29.40	55.19	30.85	45.96	22.31
NORMALIZED GLUTEUS MEDIUS MEAN EMG	203.85	47.63	67.59	35.39	66.09	36.80
NORMALIZED ERECTOR SPINAE MEAN EMG	122.12	112.44	124.92	180.68	136.65	190.11
NORMALIZED RECTUS FEMORIS MEAN EMG	405.72	274.64	277.71	156.49	318.21	203.85
NORMALIZED MAXIMUM VERTICAL GRF	154.11	33.05	153.89	23.93	163.23	24.68

All EMG values are expressed as a percentage of MVC. All GRF values are expressed as a percentage of bodyweight. *Significant difference between conditions (*p*<.05).

Table 3

Lateral Cutting Task-1a

BRACE TYPE	NO BRACE		ASO EVO		AAT1	
	MEAN	SD	MEAN	SD	MEAN	SD
NORMALIZED PERONEUS LONGUS MEAN EMG	247.95	201.94	239.98	225.46	208.16	208.45
NORMALIZED L. GASTROCNEMIUS MEAN EMG	159.12	94.09	189.74	154.21	259.65	351.99
NORMALIZED BICEPS FEMORIS MEAN EMG	137.33	65.05	99.07	41.84	136.62	81.93
NORMALIZED GLUTEUS MEDIUS MEAN EMG	684.23	1217.94	461.80	947.21	670.13	1361.26
NORMALIZED ERECTOR SPINAE MEAN EMG	179.08	246.40	511.37	1045.41	502.47	746.94
NORMALIZED RECTUS FEMORIS MEAN EMG*	583.57	590.83	259.47	128.28	195.51	98.28
NORMALIZED MAXIMUM VERTICAL GRF	126.14	8.91	125.64	10.18	127.49	9.70
NORMALIZED MAXIMUM MEDIOLATERAL GRF*	22.43	8.37	26.16	7.06	29.55	10.26
NORMALIZED MAXIMUM ANTEROPOSTERIOR GRF	50.11	9.91	39.62	17.60	54.13	15.84

All EMG values are expressed as a percentage of MVC. All GRF values are expressed as a percentage of bodyweight. *Significant difference between conditions ($p < .05$).

Table 4

Lateral Cutting Task-1b

BRACE TYPE	NO BRACE		ASO EVO		AAT1	
	MEAN	SD	MEAN	SD	MEAN	SD
NORMALIZED PERONEUS LONGUS MEAN EMG	188.12	106.52	210.11	162.27	173.97	103.54
NORMALIZED L. GASTROCNEMIUS MEAN EMG	162.38	108.23	162.60	96.28	221.68	163.30
NORMALIZED BICEPS FEMORIS MEAN EMG	98.80	38.32	95.18	40.92	99.13	42.00
NORMALIZED GLUTEUS MEDIUS MEAN EMG	423.07	576.11	250.53	332.31	464.58	879.70
NORMALIZED ERECTOR SPINAE MEAN EMG	144.72	108.40	287.57	410.59	226.16	201.46
NORMALIZED RECTUS FEMORIS MEAN EMG	510.55	718.26	268.28	144.97	265.51	123.17
NORMALIZED MAXIMUM VERTICAL GRF	126.14	8.91	125.63	10.18	127.49	9.70
NORMALIZED MAXIMUM MEDIOLATERAL GRF	58.72	6.88	52.79	14.62	59.57	11.23
NORMALIZED MAXIMUM ANTEROPOSTERIOR GRF	50.11	9.91	39.62	17.60	54.13	15.84

All EMG values are expressed as a percentage of MVC. All GRF values are expressed as a percentage of bodyweight. *Significant difference between conditions ($p < .05$).

Table 5

Lateral Cutting Task-2a

BRACE TYPE	NO BRACE		ASO EVO		AAT1	
	MEAN	SD	MEAN	SD	MEAN	SD
NORMALIZED PERONEUS LONGUS MEAN EMG	184.66	116.55	245.74	286.77	222.64	2636
NORMALIZED L. GASTROCNEMIUS MEAN EMG	289.35	214.55	246.55	169.01	446.55	706.77
NORMALIZED BICEPS FEMORIS MEAN EMG	105.18	53.34	88.35	44.94	107.04	71.35
NORMALIZED GLUTEUS MEDIUS MEAN EMG*	339.53	476.26	179.54	198.48	587.09	1228.60
NORMALIZED ERECTOR SPINAE MEAN EMG	297.24	498.40	750.83	1564.35	474.33	786.34
NORMALIZED RECTUS FEMORIS MEAN EMG	236.87	182.24	153.70	47.10	139.47	47.44
NORMALIZED MAXIMUM VERTICAL GRF	116.06	18.81	119.21	19.32	117.30	10.99
NORMALIZED MAXIMUM MEDIOLATERAL GRF	52.16	13.59	55.00	10.11	54.98	11.56
NORMALIZED MAXIMUM ANTEROPOSTERIOR GRF	1.92	2.67	1.38	2.34	2.38	3.84

All EMG values are expressed as a percentage of MVC. All GRF values are expressed as a percentage of bodyweight. *Significant difference between conditions ($p < .05$).

Table 6

Lateral Cutting Task-2b

BRACE TYPE	NO BRACE		ASO		T1	
	MEAN	SD	MEAN	SD	MEAN	SD
NORMALIZED PERONEUS LONGUS MEAN EMG	216	129.78	204.39	142.76	189.27	105.94
NORMALIZED L. GASTROCNEMIUS MEAN EMG	219.17	155.25	2095	121.68	2782	262.60
NORMALIZED BICEPS FEMORIS MEAN EMG	95.15	36.74	83.65	32.04	101.53	48.58
NORMALIZED GLUTEUS MEDIUS MEAN EMG	247.82	251.11	166.31	106.32	611.83	1076.21
NORMALIZED ERECTOR SPINAE MEAN EMG	129.54	103.51	172.25	135.50	189.67	243.94
NORMALIZED RECTUS FEMORIS MEAN EMG	411.12	465.11	246.14	98.08	250.84	95.30
NORMALIZED MAXIMUM VERTICAL GRF	136.11	12.49	137.04	12.67	137.96	14.34
NORMALIZED MAXIMUM MEDIOLATERAL GRF	52.16	13.59	55.00	10.11	54.98	11.56
NORMALIZED MAXIMUM ANTEROPOSTERIOR GRF	2.43	2.82	2.98	1.16	3.12	1.65

All EMG values are expressed as a percentage of MVC. All GRF values are expressed as a percentage of bodyweight. *Significant difference between conditions ($p < .05$).

Inferential Statistics.

Vertical jump take off. There was a significant mean difference in normalized LG mean EMG activity during vertical jump take off, depending on the bracing condition, X^2 (2) = 7.400, p = .025 (Figure 7). Post hoc analysis revealed no significant differences in normalized LG mean EMG activity when wearing the AE ankle braces (M =191.76, SD = 92.26) compared to the no brace control condition (M = 209.86, SD = 102.81; Z = -2.293, p = .022). There was no significant differences in normalized LG mean EMG activity when wearing the T1 ankle braces (M = 201.62, SD = 94.9) compared to the no brace control condition (Z = -1.580, p = .114). There was no significant differences in normalized LG mean EMG activity when wearing the AE ankle braces compared to T1 ankle braces (Z = -.459, p = .646).

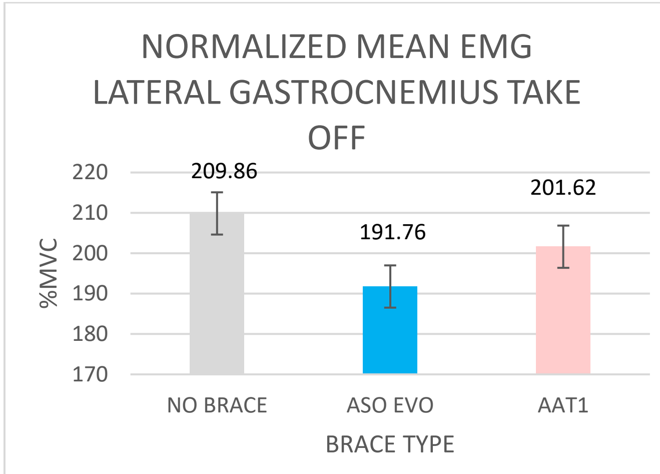


Figure 7. Normalized Mean EMG Activity Lateral Gastrocnemius Vertical Jump Take Off. This figure illustrates differences across brace types with respect to normalized mean EMG activity of the lateral gastrocnemius muscle during vertical jump take off.

Lateral cutting task-1a. There was a significant mean difference in normalized RF mean EMG activity during the lateral cutting task (1a) depending on the bracing condition, X^2 (2) = 9.314, p = .009 (Figure 8). Post hoc analysis revealed no significant differences in normalized RF mean EMG activity when wearing the AE ankle braces (M = 259.47, SD = 128.28) compared to the no brace control condition (M = 583.57, SD = 98.28; Z = -2.073, p = .038). There was no significant difference in normalized RF mean EMG activity when wearing the T1 ankle braces (M = 195.51, SD = 98.28) compared to the AE ankle braces (Z = -1.680, p = .093). There was a significant reduction in normalized RF mean EMG activity between the no brace control condition and the T1 ankle braces (Z = -2.429, p = .015) with a large effect size (r = .5).

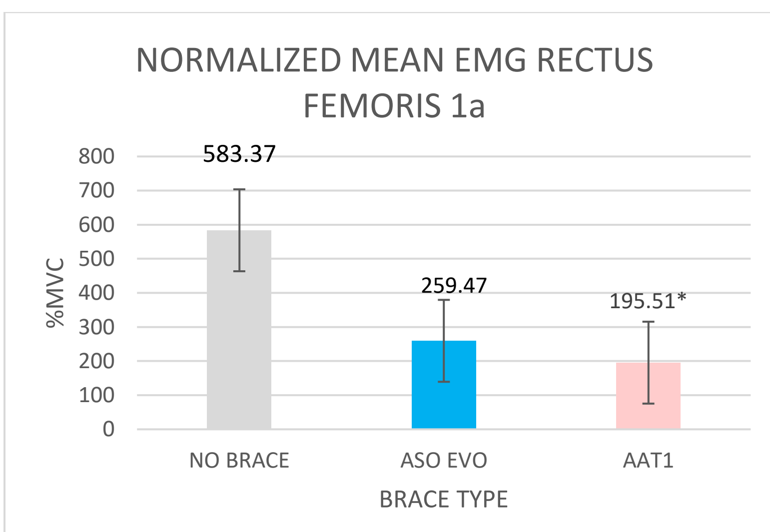


Figure 8. Normalized Mean EMG Activity Rectus Femoris 1a. This figure illustrates differences across brace types with respect to normalized mean EMG activity of the rectus femoris muscle during the lateral cutting task (1a).*Significant difference from no brace at $p = .05$.

There was a significant mean difference in maximum mediolateral GRF during the lateral cutting task (1a) depending on the brace condition, $X^2 (2) = 8.359$, $p = .015$ (Figure 9). Post hoc analysis revealed no significant differences in maximum mediolateral GRF when wearing the AE ankle braces (M = 26.16. SD = 7.06) compared to the no brace control condition (M = 22.43, SD = 8.37; $Z = -1.988$, $p = .047$). There was no significant differences in maximum mediolateral GRF when wearing the T1 ankle braces (M = 29.55, SD = 10.26) compared to the no brace control condition ($Z = -2.090$, $p = .037$). There was no significant differences in maximum mediolateral GRF when wearing the T1 ankle braces compared to the AE ankle braces ($Z = -1.244$, $p = .214$).

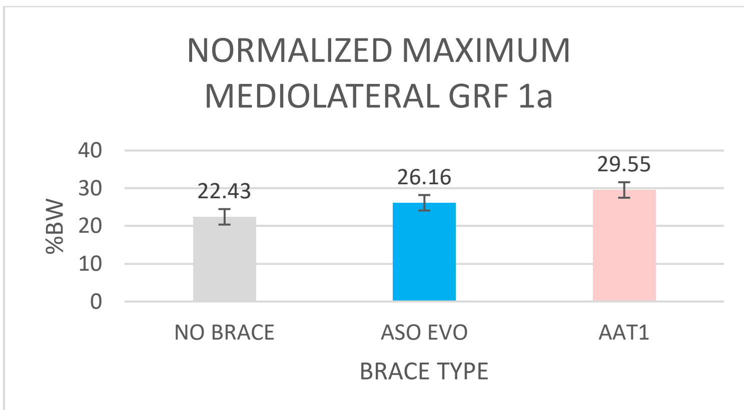


Figure 9. Normalized Maximum Mediolateral GRF 1a. This figure illustrates differences across brace types with respect to mean normalized maximum mediolateral GRF values during the lateral cutting task 1a.

Lateral cutting task-2a. There was a significant mean difference in normalized GM mean EMG activity during the lateral cutting task (2a) depending on the brace condition, $X^2 (2) = 7.774$, $p = .021$ (Figure 10). Post hoc analysis revealed no significant differences in normalized GM mean EMG activity when wearing the AE ankle braces (M = 179.54, SD = 198.48) compared to the no brace control condition (M = 339.53 SD = 476.26; $Z = -1.988$, $p = .047$). There was no significant differences in normalized GM mean EMG activity when wearing the T1 ankle braces (M = 587.09, SD = 1228.60) compared to the no brace control condition ($Z = -.968$, $p = .333$). There was no significant differences in normalized GM mean EMG activity when wearing T1 ankle braces compared to AE ankle braces ($Z = -.415$, $p = .678$).

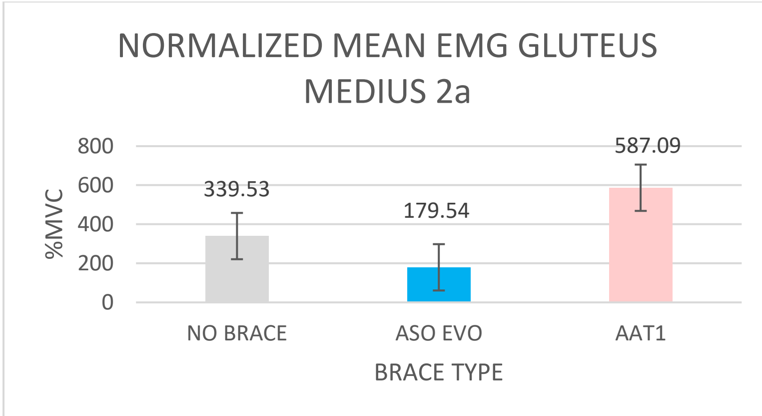


Figure 10. Normalized Mean EMG Activity Gluteus Medius 2a. This figure illustrates differences across brace types with respect to normalized mean EMG activity of the gluteus medius during 2a. * Significant difference at p = .05 only.

Discussion

The primary purpose of this preliminary study was to determine if the protocol and tasks were feasible to detect changes in the biomechanics of the lower extremity and lumbar spine when wearing ankle braces. Specifically, if differences in lower extremity and lumbar spine EMG muscular activity, GRFs, and performance could be detected during a Vertical Jump Test and lateral cutting task when wearing the AE softshell and T1 semi-rigid ankle braces compared to no bracing. No significant differences in vertical jump height or lateral cutting task time between brace conditions was revealed. With respect to the lateral cutting task, a significant reduction in RF EMG activity during the lateral cutting task (1a) was revealed when wearing T1 ankle braces. The Friedman Test also revealed a significant difference between conditions in normalized LG mean EMG activity during vertical jump take off, normalized maximum mediolateral GRF during the lateral cutting task (1a), and mean GM EMG activity during the lateral cutting task (2a). When a Bonferroni correction was applied, post hoc analysis did not reveal a significant difference between conditions for these variables.

With respect to vertical jump height, the results of this study somewhat contradict that of Henderson et al. (2016), Parsley et al. (2013), and Smith et al. (2016), despite using a similar methodology. Henderson et al. found a significant reduction in vertical jump height when wearing the T1 ankle braces. Additionally, Smith et al. found a significant reduction in vertical jump height when wearing the McDavid 195T™ ankle braces. Furthermore, Parsley et al. (2013) noted a significant reduction in vertical jump height when wearing the AE and Aircast Airsport™ ankle braces. While a significant reduction in vertical jump height was not revealed in this study, there was a trend towards an overall reduction in vertical jump height when wearing ankle braces. On average, vertical jump height was .68 cm lower than the no brace condition when wearing the AE ankle braces, and 1.87 cm lower when wearing the T1 ankle braces. Given this study's small sample size and the results of recent studies, it is possible that these differences would have been significant in a larger sample. It is also possible, however, that these differences are a result of study design; participants performed the vertical jump on the same day, in the same order in which the reduction in vertical jump height was observed. Thus, decreases in vertical jump height may be the result of participant fatigue rather than brace condition.

The observed decrease in LG EMG activity when wearing the AE ankle braces is somewhat in line with previous research. Smith et al. (2016) observed a significant decrease in soleus EMG activity during a vertical jump when wearing the McDavid 195T™ ankle braces. Additionally, a decrease that approached significance (p = .06) was observed in the gastrocnemius muscle EMG activity. As such, the reduction in LG EMG activity during take off observed in this study is not surprising. While not the same muscle, both the gastrocnemius and soleus insert on the calcaneus and contribute to plantarflexion of the foot (Tortora & Nielson, 2010). Therefore, the reduction in dorsiflexion and plantarflexion ROM by an ankle brace (DiStefano et al., 2008; Gudibanda & Wang, 2005) is the likely cause of reduced plantarflexor EMG activity (Smith et al., 2016). Furthermore, Smith et al. (2016) suggested that this reduction in ankle ROM and soleus activity contributed to the decreased vertical jump height; however, a significant reduction in vertical jump height was not observed in this study.

Smith et al. (2016) suggested that the reduced ankle ROM and soleus muscle EMG activity resulted in the reduction in vertical jump height observed in their study. Although an important factor, ankle angle at take off only accounts for 21% of the variance in the vertical jump performance (Aragón-Vargas & Gross, 1997). In comparison, models that included hip power, hip torque, and knee extension strength accounted for up to 60% of the explained variance in vertical jump performance (Aragón-Vargas & Gross, 1997). During vertical jumping, the RF muscle is responsible for both knee extension and hip flexion (Jacobs, Bobbert, & van Ingen Schenau, 1996; Tortora & Nielson, 2010). Though not significant, a decrease of 44.5% and 47.47% in RF EMG activity during the Vertical Jump Test take off was recorded in this study when wearing AE and T1 ankle braces, respectively. Because EMG muscular activity has been linearly associated with force production (Onishi et al., 2000), a reduction in RF EMG activity may indicate a reduction in knee extension force and hip power, resulting in a decreased vertical jump height. To a degree, this was the case in this study.

No difference in time to complete the lateral cutting task was observed between conditions. The lateral cutting task was unique to this study, therefore, it is difficult to make direct comparison to studies that have used timed agility tests. The results are in line with Henderson et al. (2016) who used the T-test Agility Test, in which the lateral cutting task in this study is partially based on. Interestingly, there was a relatively large reduction in RF EMG muscular activity during the lateral cutting task (1a) when wearing the ASO EVO ™ ankle braces. Furthermore, a significant reduction in RF EMG activity was revealed when wearing the T1 ankle braces. The reduction of RF EMG activity when ankle ROM is restricted during an eccentric movement is comparable to the findings reported by Macrum et al. (2012). Vastus medialis and vastus lateralis muscles EMG activity decreased during the squat when ankle plantarflexion was artificially restricted. Thus, it appears that altering kinematics at the ankle during functional tasks can affect EMG activity above the ankle, specifically in the quadriceps muscles. As this study did not examine kinematic variables, it is unknown if the reduction in RF EMG activity when wearing the ankle braces is due to altered knee and/or hip kinematics, or some other mechanism.

Maximum vertical GRF during the Vertical Jump Test was not significantly affected in this study by wearing the AE ankle braces or T1 ankle braces. In agreement with DiStefano et al.

(2008), mean values for maximum vertical GRF in the no brace control and the AE conditions differed by less than 1 N. In comparison, though not significant, the T1 ankle braces produced vertical GRF approximately 10 N greater at landing than the no brace control and AE conditions. This increase in vertical GRF when wearing semi-rigid ankle braces is somewhat in line with Hodgson et al. (2005). Hodgson et al. observed a significant increase in vertical GRF during a drop landing when wearing the Active Ankle T2™ ankle braces. Due to the difference in methodology, Hodgson et al. standardized the drop height. When wearing the T1 ankle braces participants produced slightly smaller vertical jump heights in this study, it is hypothesized then that the vertical GRF at landing should have been lower, not higher. Therefore, further evaluation is required as a significant difference may have been detected in a larger sample.

Using a similar cutting task, Klem et al. (2016) observed a reduction in anterior shear forces at the knee when wearing the Active Ankle T2™ ankle braces. Large anterior shear force at the knee is a by-product of forceful quadriceps contraction (DeMorat, Weinhold, Blackburn, Chudik, & Garret, 2004). Excessive anterior shear forces as the result of forceful quadriceps contraction has been associated with injuries to the ACL (DeMorat et al., 2004; Koga et al., 2016). Thus, the reduction observed in RF muscle EMG activity during the lateral cutting task (1a) when wearing any type of ankle brace may indicate a reduction in anterior shear force at the knee. Klem et al. (2016) also noted that softshell and semi-rigid ankle braces increased knee internal rotation ROM and abduction angles during a 90 degree cutting task. Increased internal and external rotation ROM at the knee joint (Noyes, Mooar, Matthews, & Butler, 1983) and knee abduction angle (Hewett, Torg, & Boden, 2009) has also been associated with ACL injuries. While kinematic data were not collected in this study, an increase in mediolateral GRF was observed when wearing a semi-rigid ankle brace. As the lateral cutting task (1a) should involve movement primarily in the sagittal plane, this could suggest changes in knee abduction/adduction angle and knee internal/external rotation moments as force is directed towards the ground more laterally. Thus, the potential benefits of reducing quadriceps activation and anterior shear force at the knee during a 90 degree cutting task may be negated.

Though not significant, a reduction in GM muscle EMG activity was observed when wearing the AE ankle braces compared to the no brace control condition. Additionally, there was an increase in GM muscle EMG activity when wearing the T1 ankle braces compared to the no brace control condition. As the brace conditions appeared to affect GM muscle EMG activity in the opposite manner, the differences may due to brace design. The AE ankle brace should allow slightly more motion laterally than the T1, due to the lack of rigid sides. Therefore, the T1 ankle brace may produce more shear forces when moving laterally, as the fibula is pushed against the rigid sides and the ankle is prevented from inverting. As the GM muscle functions to abduct and externally rotate the hip (Tortora & Nielson, 2010), the increase in GM muscle EMG activity may indicate a compensation strategy for the reduced ROM at the ankle. While this may explain the increase in GM muscle EMG activity when wearing a semi-rigid ankle brace, it does not explain the decrease observed when wearing the AE ankle braces.

Studies that have found performance and kinematic changes in the lower leg have attributed the results to the restriction in ankle ROM when wearing ankle braces but there may be a neuromuscular explanation. The AE is a softshell lace-up ankle brace with horseshoe and heel lock strap design (Gudibanda & Wang, 2005) providing more skin contact around the ankle than the T1 ankle brace. Motor programming during agility tasks relies on central nervous system (CNS) input and afferent feedback from the involved distal structures (Craig, 2004). When wearing ankle braces, it is thought that cutaneous receptors in the contacted skin become more active, increasing afferent feedback to the CNS (Feuerbach, Grabiner, Koh, & Weiker, 1994). During gait, a non-noxious cutaneous stimuli, which an ankle brace would provide, to the tibial and sural nerves has been shown to have an inhibitory response in RF muscle when transitioning from the swing to the stance phases (Yang & Stein, 1990). Furthermore, Bullock-Saxton, Janda, and Bullock (1994) noted a delayed activation of the gluteus maximus muscle during hip extension in subjects with a previous history of unilateral ankle sprains. As such, they suggested that alterations in afferent feedback, stemming from an ankle sprain, can influence motor programing above the ankle. As gluteus medius activity was reduced in the AE condition, but increased in T1 condition during the lateral cutting task (2a), it is possible that differences in afferent feedback provided by each ankle brace may alter motor programs utilized for locomotion.

Limitations

As the sample size of this pilot study was low (n = 10), the results have limited clinical application. Furthermore, the non-parametric nature of the Friedman Test limits the power of any findings. The Friedman Test revealed significant differences between conditions for four variables. When a post-hoc analysis was performed and a Bonferroni correction applied, however, only one variable reached significance The purpose of a Bonferroni correction is to reduce the chances of a Type 1 error when conducting multiple pairwise comparisons (Armstrong, 2014). As such, a lower p value was used ($p = .017$) to determine significance when performing post hoc analysis. Although post hoc analysis produced p values $<.05$, p values were not less than .017. Therefore, given the nonparametric nature of the Friedman Test and low sample size (n = 10), it is possible that the Friedman Test produced a type 1 error for these variables.

Some participants had difficulty selectively activating the intended muscles when performing the MVCs. As a result, this could have affected the results. In terms of methodology for the Vertical Jump Test, participants took off and landed with only one foot on the force platform. Thus, GRFs are only representative of the participant's dominant extremity. Participants were instructed to take off and land evenly on both feet; however, if participants favoured one side when taking off, this could have influenced GRF data during the Vertical Jump Test. During the lateral cutting task, some participants noted that there was not enough traction on or around the force platform, and were sliding when contacting with the force platform. As such, this could have affected GRF during the lateral cutting task. Electromyography may have also been affected if gait was altered to compensate for the sliding. None of the participants in the study had previously worn ankle braces. Therefore, results may be different in persons who have experience wearing ankle braces.

Conclusions

The methodology revealed significant differences and trends in EMG and GRF when comparing the no brace controlled condition to the braced conditions. Furthermore, EMG activity of the lower leg musculature during the vertical jump take off and lateral cutting task (1a) was in line with previous literature. Muscle activation patterns during the lateral cutting task were also similar to those observed by Houck (2003). This study provides the basis for future research, using a similar methodology, on the relationship between lower extremity and lumbar spine EMG muscle activity, GRF, and performance measures when wearing ankle braces. The addition of a non-slip surface on and around the force platform should be implemented in future studies. In addition to using a larger sample size, future studies should investigate EMG activity, GRFs, and performance when wearing ankle braces in populations with experience wearing ankle braces or pathological and injured populations.

References

Ambegaonkar, J. P., Redmond, C. J., Winter, C., Ambegaonkar, S. J., Thompson, B., & Guyer, S. M. (2011). Ankle stabilizers affect agility but not vertical jump or dynamic balance performance. *Foot and Ankle Specialist, 4*, 354-36.

Aragón-Vargas, L. F., & Gross, M. M. (1997). Kinesiological factors in vertical jump performance: Differences among individuals. *Journal of Applied Biomechanics, 13*, 24-44.

Armstrong, R. A. (2014). When to use the Bonferroni correction. *Ophthalmic and Physiological Optics, 34.* 502-508. doi: 10.1111/opo.12131

Craig, B. W. (2004). What is the scientific basis of speed an agility? *Strength and Conditioning Journal, 26*(3), 13-14.

CSEP. (2013). Canadian society for exercise physiology- physical activity training for health. Ottawa, Canada: Canadian Society for Exercise Physiology.

Delsys Inc. (2012). *Delsys Trigno Wireless system: User's guide*. Boston, MA: Delsys Inc.

DeMorat, G., Weinhold, P., Blackburn, T., Chudik, S., & Garret, W. (2004). Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *American Journal of Sports Medicine, 32*, 477-483. doi: 10.1177/0363546503258928

Devita, P., & Skelly, W. A. (1992). Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Medicine and Science in Sports and Exercise, 24,* 108-115.

Dufek, J. S., & Bates, B. T. (1990). The evaluation and prediction of impact forces during landings. *Medicine and Science in Sports and Exercise, 22*, 370-377.

DiStefano, L. J. Padua, D. A., Brown, C. N., & Guskiewicz, K. M. (2008). Lower extremity kinematics and ground reaction forces after prophylactic lace-up ankle bracing. *Journal of Athletic Training, 43*, 234-241.

Ferran, N. A., Maffulli, N. (2006). Epidemiology of sprains of the lateral ankle ligament complex. *Foot and Ankle Clinics, 11*, 659-662. http://dx.doi.org/11016/j.fcl.2006.07.002

Feuerbach, J. W., Grabiner, M. D., Koh, T. J., & Weiker, G. G. (1994). Effect of an ankle orthosis and ankle ligament anesthesia on ankle joint proprioception. American Journal of Sports Medicine, 22, 223-229. doi: 10.1177/036354659402200212

Gross, T. S., & Nelson, R. C. (1988). The shock attenuation role of the ankle during landing from a vertical jump. *Medicine and Science in Sports and Exercise, 20*, 506-514.

Gudibanda, A., & Wang, Y. (2005). Effect of the ankle stabilizing orthosis on foot and ankle kinematics during cutting maneuvers. *Research in Sports Medicine, 13*(2), 111-126. doi:11080/15438620590956115

Hewett, T. E., Torg, J. S., & Boden, B. P. (2009). Video analysis of trunk and knee motion during non-contact anterior cruciate ligament injury in female athletes: Lateral trunk and knee abduction motion are combined components of the injury mechanism. *British Journal of Sports Medicine, 43*, 417-422.

Hodgson, B., Tis, L., Cobb, S., & Higbie, E. (2005). The effect of external ankle support on vertical ground-reaction force and lower body kinematics. *Journal of Sport Rehabilitation, 14*, 301-312. http://dx.doi.org/11123/jsr.14.4.301

Houck, J. (2003). Muscle activation patterns of selected lower extremity muscles during stepping and cutting tasks. *Journal of Electromyography and Kinesiology, 13*, 545-554. doi: http://dx.doi.org/10.1016/S1050-6411(03)00056-7

Hsieh, C. T., & Cheng, L. (2016). Kinematic factors in countermovement jump for female volleyball players with different skill levels. *International Journal of Sports Science, 6*, 6-1 doi: 15923/j.sports.20160601.02

Jacobs, R., Bobbert, M. F., & van Ingen Schenau, G. J. (1996). Mechanical output from individual muscles during explosive leg extensions: The role of biarticular muscles. *Journal of Biomechanics, 29*, 513-523.

Kaminski, T.W., Hertel, J., Amendoloa, N., Docherty, C. L., Dolan, M. G. Hopkins, J. T., Nussbaum, E., Poppy, W., & Richie, D. (2013). National athletic trainers association position statement: Conservative management and prevention of ankle sprains in athletes. *Journal of Athletic Training, 48*, 528-545. doi: 14085/1062-6050-48.02

Klem, N-R., Wild, C. Y., Williams, S. A., & Ng, L. (2016). Effect of external ankle support on ankle and knee biomechanics during the cutting maneuver in basketball players. *American Journal of Sports Medicine*. Advance online publication. doi: 11177/0363546516673988

Koga, H., Muneta, T., Bahr, R., Engebretsen, L., & Krosshaug, T. (2016). ACL injury mechanisms: Lessons learned from video analysis. In V. Musahl, J. Karlsson, R. Kuroda, & S. Zaffagnini (Eds.). *Rotary knee instability: An evidence based approach* (27-36). Switzerland: Springer.

Leppänen, M., Aaltonen, S., Parkkari, J., Heinonen, A., & Kujala, U. M. (2014). Interventions to prevent sports related injuries: A systematic review and meta-analysis of randomized controlled trials. *Sports Medicine,44*, 473-486. doi: 11007/s40279-013-0136-8

Macrum, E., Bell, D. R., Boling, M., Lewek, M., & Padua, D. (2012). Effect of limiting ankle-dorsiflexion range of motion on lower extremity kinematics and muscle-activation patterns during a squat. *Journal of Sport Rehabilitation*, *21*, 144-150.

Nilsson, J., & Thorstensson, A. (1989). Ground reaction forces at different speeds of human walking and running. *Acta Physiologica, 136*, 217-227. doi: 10.1111/j.1748-1716.1989.tb08655.x

Noyes, F. R., Mooar, P. A., Matthews, D. S., & Butler, D. L. (1983). The symptomatic anterior cruciate-deficient knee. Part 1: The long-term functional disability in athletically active individuals. *Journal of Bone and Joint Surgery, 65*, 154-162.

Onishi, H., Yagi, R., Akasaka, K., Momose, K., Ihashi, K., & Yanda, Y. (2000). Relationship between emg signals and force in human vastus lateralis muscle using multiple bipolar wire electrodes. *Journal of Electromyography and Kinesiology, 10*, 59-67. https://doi.org/10.1016/S1050-6411(99)00020-6

Parsley, A., Chinn, L., Lee, S. Y., Ingersoll, C., & Hertel, J. (2013). Effect of 3 different ankle braces on functional performance and ankle range of motion. *Athletic Training and Sports Health Care, 5*(2), 69-75.

Petersen, W., Rembitzki, I. V., Koppenburg, A. G., Ellermann, A., Liebau, C., Brüggemann. G. P., & Best, R. (2013). Treatment of acute ligament injuries: A systematic review. *Arthroscopy and Sports Medicine, 133*, 1129-1141. doi: 11007/s00402-013-1742-5

Roos, K. G., Kerr, Z. Y., & Mauntel, T. C. (2016). The epidemiology of lateral ligament complex ankle sprains in national collegiate athletic association sports. *American Journal of Sports Medicine, 45*, 201-208. http://dx.doi.org/11177%2F0363546516660980

SENIAM. (2014). *Surface electromyography for the non-invasive assessment of muscles*. Retrieved from http://seniam.org/

Smith, B., Claiborne, T., & Liberi, V. (2016). Ankle bracing decreases vertical jump height and alters lower extremity kinematics. *International Journal of Athletic Therapy and Training, 21*(2), 39-46. https://doi.org/11123/ijatt.2014-0143

Tortora, G. J., & Nielson, M. (2010). *Principles of human anatomy* (4th ed.). Jefferson City, U.S.A: John Wiley and Sons, Inc.

Verhagen, E. A. L. M., & Bay, K. (2010). Optimising ankle sprain prevention: a critical review and practical appraisal of the literature. *British Journal of Sports Medicine, 44*, 1082-1088. doi: 11136/bjsm.201076406

Wernli, K., Ng, L., Phan, X., Davey, P., & Grisbrook, T. (2016). The relationship between landing sound, vertical ground reaction force, and kinematics of the lower limb during drop landings in healthy men. *Journal of Orthopaedic and Sports Physical Therapy, 46*, 194-199. doi: 10.2519/jospt.2016.6041

West, T., Ng, L., & Campbell., A. (2013). The effect of ankle bracing on knee kinetics and kinematics during volleyball specific tasks. *Scandinavian Journal of Medicine and Science in Sports, 24*, 958-963. doi: 11111/sms.12130

Yang, J. F., & Stein, R. B. (1990). Phase-dependent reflex reversal in human leg muscles during walking. *Journal of Neurophysiology, 63*, 1109-1117.