Evaluation of biomechanical and neuromuscular effects of prophylactic knee brace use following exercise

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners. I understand that my thesis may be made electronically available to the public.

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Abstract

The use of knee braces prophylactically is still considered as an approach for injury mitigation for those in high-risk sporting activities, though their use is not fully supported. The purpose of this thesis was to examine biomechanical and neuromuscular effects of prophylactic brace wear following standardized repetitive exercise. Twelve participants participated and acted as their own control. The participants were required to participate in two sessions, one control session with no brace and one intervention session with the application of a off-the-shelf prophylactic knee brace. Pre-and post-exercise intervention single leg drop landings were recorded to examine the effects of an acute exercise stimulus on the neuromuscular and biomechanical effects of brace wear. Additionally, trials were collected at 30-minutes post-exercise to examine residual effects of the brace wear on landing kinematics and kinetics. Difference tests using analysis of variance (ANOVA) showed that there was a minimal effect of the prophylactic knee brace on biomechanical and neuromuscular variables following exercise as well as 30-minutes following knee brace removal. Further research may be required to identify if braces can be worn prophylactically to reduce the risk of injury during activity.

Keywords: Anterior Cruciate Ligament; Bracing; Injury Prevention; Brace Adaptations, Exercise, Biomechanical Changes; Neuromuscular Effects

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Chapter 1: Thesis Overview

Incidence of ACL injury is associated with active populations, poor rehabilitation status, and increased risk of secondary injuries (Finsterbush et al., 1990; Lustosa et al., 2011). Patients with ACL injuries are projected to have approximately a 50% chance of developing knee osteoarthritis in the ipsilateral knee within 15 years after the initial injury (Shimokochi and Shultz, 2008). This illustrates the importance of injury prevention for the knee joint to mitigate post-ACL injury issues related to returning to original function and preventing potential future musculoskeletal complications.

There are several proposed strategies for reducing the risk of ACL injury in high-risk individuals. These strategies can include neuromuscular jump training and/or prophylactic bracing for individuals who are classified as at-risk group for ACL injury (Baltaci et al., 2011; Hewett et al., 2005). Evidence has suggested that neuromuscular jump training is a preventative strategy for at-risk individuals as biomechanical studies have shown significant improvement in injury rates in intervention compared to control groups (Hewett et al., 1999). Prophylactic bracing for the knee, on the other hand, is equivocal in the literature with no clear indication of its effectiveness in the prevention of injury. There remain many peripheral areas of research within the prevention of ACL injury that need to be addressed to determine whether bracing prophylactically can prevent knee injuries.

An area of prophylactic bracing that has yet to be investigated concerns the effects of removing the knee brace, and if any of the observed effects with brace application last after the removal of the brace. There were two major implications that may have presented with the findings of this work. First, if residual effects were observed after brace removal that were considered protective of the knee structures, further research could be conducted to examine if a brace could be used as an adjunct in pre-game warm-ups followed by removal of the brace for game play thereby allowing the effects to be experienced without wearing the brace. Alternatively, if the findings of this work suggested that residual effects are detrimental to the knee, caution may be warranted to the brace wearer that injury upon brace removal may be possible up to 30-minutes post-exercise. This thesis work was the first step toward answering some of these questions. The purpose of this work was to characterize potential biomechanical and neuromuscular effects of brace wear prior-to, immediately following, and 30-miuntes post-standardized exercise. It was hypothesized that detrimental changes would be observed following exercise and remain upon brace removal and again 30-minutes post-exercise.

A within subjects design was used to examine the effect of acute brace wear on biomechanical and neuromuscular variables. Participants were required to complete five single-leg landings at 5 time points throughout the session: two prior to a standardized exercise protocol (time 1 = no brace and time 2 = brace application [only in the braced intervention session]) and three proceeding after exercise (time 3 = following exercise, time 4 = brace removal [only in the braced intervention session], and time 5 = 30-minutes of rest post-brace removal). The no brace session acted as a control while the session involving brace application at the time points immediately before, during, and immediately after exercise acted as the intervention. A full set-up involving kinematics, kinetics, and electromyography was used to collect intended dependent variables.

The specific hypotheses tested with this thesis were:

1) A prophylactic knee brace will have <u>effects</u> on biomechanical and neuromuscular variables before and after 30-minutes of standardized treadmill exercise. Specifically:

a. **Peak vertical ground reaction force (vGRF) and time to peak vGRF** will decrease in the braced versus unbraced trials (Rishiraj et al., 2012)

b. **Sagittal ankle, knee, and hip angles**, as well as **knee valgus angles** at **ground contact and peak vGRF**, will be *larger* in unbraced versus braced trials (i.e., more erect) (Hewett et al., 2005)

c. **Sagittal ankle, knee and hip moments** will *decrease,* and **valgus knee moments** will *remain unchanged* at the moment of **peak vGRF** in unbraced versus braced trials (Singer and Lamontagne, 2008)

d. **EMG latencies** in the **hamstrings** during landing will *increase* with the braced condition (De Vita et al., 1996)

e. **EMG magnitude** in the **hamstrings** and **quadriceps** muscles will *decrease* with the braced condition (Handular et al., submitted)

2) Removing the brace <u>will not</u> have immediate <u>effects</u> on biomechanical and neuromuscular variables resulting from the braced exercise condition. Specifically:

a. **Peak Fz and time to peak Fz** will *remain decreased* in the braced trials

b. **Sagittal ankle, knee, and hip angles**, as well as **knee valgus angles at ground contact and peak vGRF**, will *remain larger* in the braced trials

c. **Sagittal ankle, knee, and hip** moments will *remain decreased*, and valgus **knee moments** will *remain unchanged* at the moment of **peak vGRF** in the braced trials

d. **EMG latencies** in the **hamstrings** during landing will *remain increased* in the braced trials

e. **EMG magnitude** in the **hamstrings** and **quadriceps** muscles will *remain decreased* in the braced trials

3) Thirty-minutes of rest after the removal of the prophylactic knee brace <u>will not</u> have an <u>effect</u> on biomechanical and neuromuscular variables. Specifically:

a. **Peak Fz and time to peak Fz** will *remain decreased* in the braced trials 30-minutes post-exercise

b. **Sagittal ankle, knee, and hip angles**, as well as **knee valgus angles at ground contact and peak vGRF**, will *remain larger* in the braced trials 30-minutes post-exercise

c. **Sagittal ankle, knee, and hip** moments will *remain decreased* and **valgus knee moments** will *remain unchanged* at the moment of **peak vGRF** in the braced trials 30-minutes post-exercise

d. **EMG latencies** in the **hamstrings** during landing will *remain increased* in the braced trials 30-minutes post-exercise

e. **EMG magnitude** in the **hamstrings** and **quadriceps** muscles will *remain decreased* in the braced trials 30-minutes post-exercise

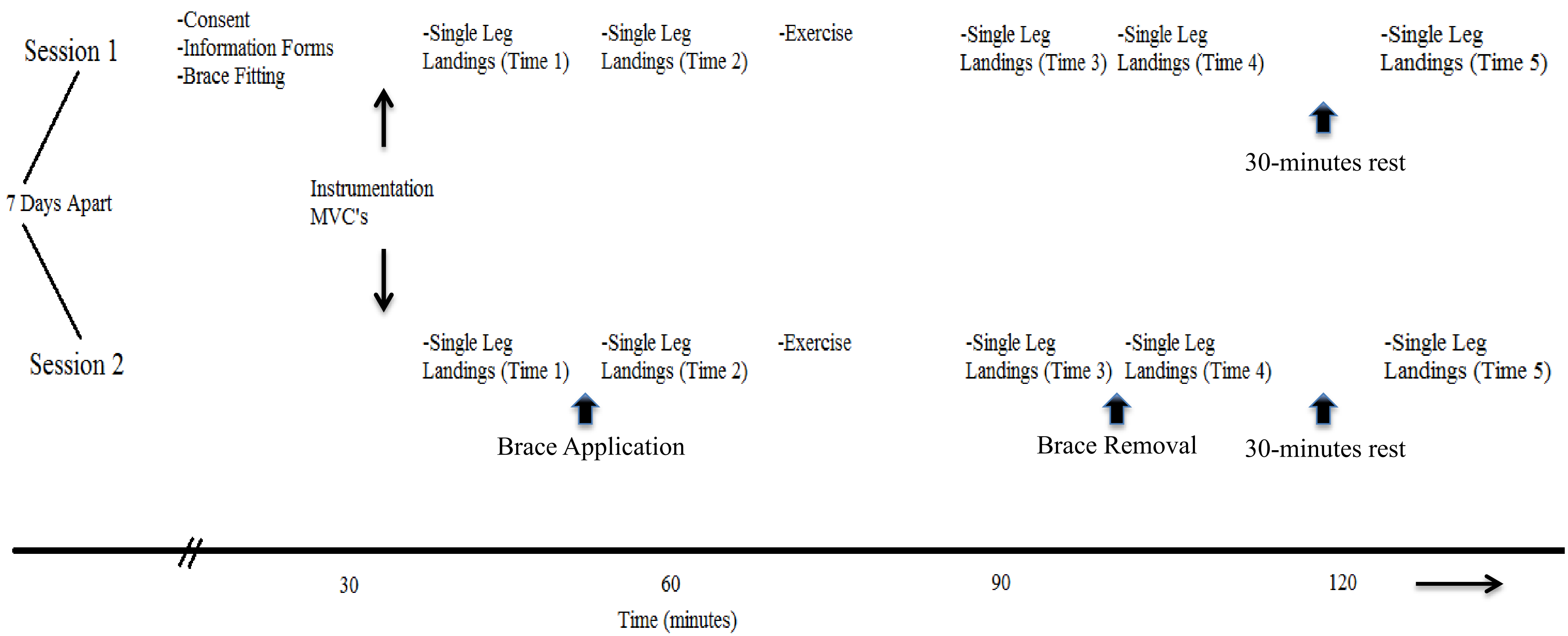

Figure 1: Schematic diagram of the experimental protocol. A within-subjects design was used to investigate the effects of the knee brace on biomechanical and neuromuscular variables.

Chapter 2: Literature Review

2.1 Knee anatomy and function

The knee consists of two articulating surfaces on the distal femur and the proximal tibia. Internally the knee contains many important structures that are important to maintain structural stability and ease of movement: numerous ligaments; the meniscus to facilitate joint conformity; cartilage to cushion the articulating surface; and bursae to lubricate the joint (Moore and Dalley 2006). These structures are passive which act to guide the knee through its passive range of motion (Goldblatt and Richmond, 2003). Externally, many tendons from neighbouring muscles serve as active constraints contributing to static and dynamic stability during weightbearing and ambulation (Shelburne et al., 2004). The concomitant actions of the passive and active structures of the knee joint provide a functional foundation for stable movement. A table outlining the major and minor passive structures of the knee is provided in Appendix A.

Movement of the knee joint through its functional range of motion and the position of the axis of rotation follows a complex set of biomechanical theories. Due to the shape of the articulating surfaces of the knee joint, describing its mechanical action as merely a sliding hinge joint would be highly simplistic (Müller 1983 p. 9). In 1836 the Weber brothers in Germany first proposed the crossed four-bar linkage theory to describe the mechanics of the tibiofemoral knee joint as a 2D planar 1- degree of freedom system. The theory marries the idea of cruciate ligament isometry with the movement of the femoral condyles on the tibial plateau in sagittal knee range of motion. The complex combination of sliding, rolling, and gliding is further compounded by motion in three dimensions during physiological loading. Figure 1 shows the Burmester curve through the

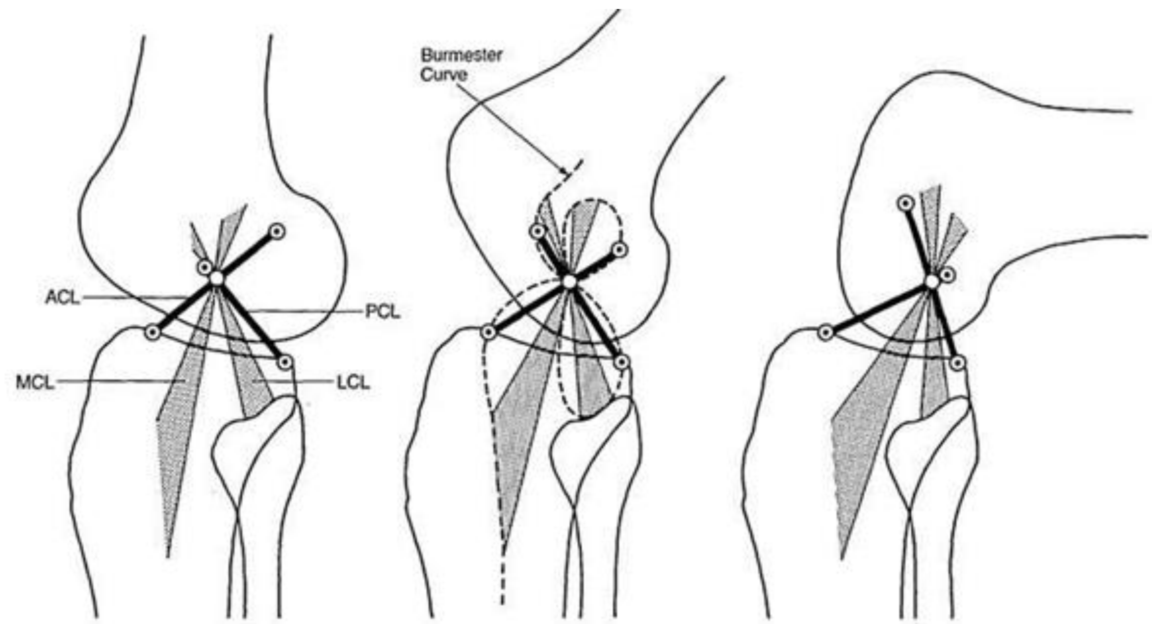

Figure 2: Crossed four-bar linkage theory of knee kinematics with subsequent Burmester Curve created by the articulation of the crossed bars. Figure from O`Brien (1992).

knee full sagittal range of motion. Therefore the mechanical geometry of knee motion is a complex system of multiple mechanical actions that must be understood to fully encapsulate the potential areas of improvement for injury prevention.

2.2 The anterior cruciate ligament (ACL)

2.2.1 Gross anatomy

The ACL runs posteroanteriorly originating on the posterior femur and attaches on the anterior aspect of the tibial plateau. Early dissection studies using cadavers have discussed the complexity of the ACL in how it functions to provide overall stability of the knee joint (Girgis et al., 1975; Fu et al., 1993; Fineberg et al., 2000). Within these studies, there is relative agreement on the existence of two separate functional bands: the anteromedial band and the posterolateral band. Additionally, some authors identify the existence of an intermediate band (Fu et al., 1993). These separate bands are considered a continuum of bands that differ in length and function with each band becoming taught at different flexion angles (Girgis et al., 1975).

2.2.2 Structure and physiology

The ACL has a unique structure that differs from other ligaments in the body. Unlike the uniform composition and organization of collagen fibres in standard ligaments, the composition and organization of collagen fibres in the ACL is variable. Strocchi et al (1992) discovered that approximately half of the collagenous fibres of the ACL were large fibres with a variable diameter throughout its fibre length while the other half were smaller fibres with a more consistent diameter throughout the fibre length. It was thought that the variable diameter fibres were responsible for resisting high tensile forces and the fibres with the consistent diameter were responsible for multivariate loading directions. The same research group found that the organization of these fibres is in a multitude of directions, and not necessarily parallel to the longitudinal axis of the ligament (Strocchi et al., 1992). The ability of the ACL to distribute high tensile forces and handle multivariate loading scenarios is important to prevent injury and maintain structural stability in the knee.

Vascular organization and natural healing processes of the ACL must be considered to understand the importance of injury prevention. The ACL's main blood supply is from the medial geniculate artery, a branch from the popliteal artery located in the popliteal fossa in the posterior aspect of the knee (Moore and Dalley 2006). Despite ample blood supply, injury to the ACL can be catastrophic from a functional point of view since the ACL rarely repairs itself fully. Unlike the MCL - which after inflammation, a reparative phase and a remodeling phase typically leads to regeneration of the completely or partially torn ligament with some structural and mechanical deficits (Frank et al., 1983) - the ACL is unable to regenerate after a complete tear (Hefti et al., 1991). Additionally, partial tears generally do not heal properly and therefore come with an increased risk of secondary injury (Hefti et al., 1991). The reason for the disparity in healing processes may be associated with decreased blood clotting in synovial joints, which typically facilitates connective tissue regeneration (Harrold 1961).

2.2.3 Function and biomechanics

The ACL serves to prevent excessive anterior tibial translation and axial rotation of the tibia with respect to the femur (Odensten and Gillquist 1985) with the anteromedial band and posterolateral band contributing to restrict these motions respectively (Yagi et al., 2002). The ACL is in a flat orientation in extension with the anteromedial band taught and proceeds to rotate 90 degrees axially during full knee flexion causing the posterolateral aspect to become taught (Odensten and Gillquist 1985; Welsh 1980). The functional implications for twisting in the longitudinal axis of the ACL allows for load sharing between the functional bands of the ACL, with *in vitro* studies showing a reciprocal action between the two bands in knee flexion where when one band becomes the taught the other will concomitantly become lax within the range of sagittal knee flexion (Girgis et al., 1975). However, Li et al (2003) showed that *in vivo*, the reciprocal action of the bands may actually occur between the ACL and PCL in weightbearing flexion instead of within the functional bands of the ACL. Nonetheless the internal ligaments of

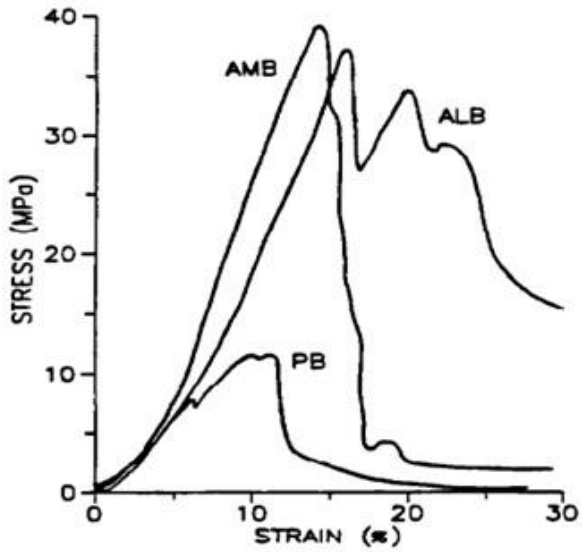

Figure 3: Stress-strain curves for the three bundles of the ACL. Note the difference in properties between the posterior and anterior bundles. Figure from Butler et al., 1992.

the knee must be considered together in their action to work in concert to keep the knee in check during knee flexion.

Understanding the loading mechanics of the ACL is important for assessing safe loading scenarios for injury prevention. Ligaments are viscoelastic, and therefore their strain and deformation properties are rate and history dependent. The non-linearity of its load-deformation curve is characterized by a toe-region where unwinding of the collagen fibres take place, a linear elastic region, and a yielding point where plastic deformation occurs (Noyes et al., 1974). Noyes et al (1974) demonstrated the rate dependent viscoelastic features of the femur-ACL-tibia complex by determining that the ACL ultimately fails at a higher absolute load, greater elongation, and absorbs more energy in a high rate loading scenario compared to a slow loading scenario. They also found that the injury outcome that predominately manifests during a fast loading scenario is a ligamentous injury while the slow rate will manifest in an avulsion of the ACL from the tibia (Noyes et al., 1979). Since we know that the ACL contains multiple bundles, biomechanical responses of each bundle with loading should be considered. In general, the slope of the elastic region and location of the yield point dramatically changes when damage is sustained in the ACL affecting the overall stiffness and ultimate strength of the ligament (Hefti et al.,1991). Figure 2 shows the results from Butler et al's (1992) study that defined stress-strain properties of the three bundles of the ACL. Accordingly, in order to reduce the negative consequences of ACL injuries, it is important to identify mechanisms causing ACL injuries, and identify what can be done to mitigate any injurious movements putting the athlete at risk for ACL injury.

2.3 ACL injuries

The ACL is one of the most commonly injured ligaments in the knee joint and injury may occur in a number of different sporting scenarios. Incidence rates for ACL injuries reported by the NCAA Injury Surveillance Report were as high as 0.33/1000 athlete-exposures for women's gymnastics and men's spring practice football, followed closely by women's soccer and women's basketball, with an average incidence of 0.15/1000 athlete-exposures across 15 NCAA sports (Hootman et al., 2007). To put this statistic into perspective, the same study demonstrated that documented concussions occurred at an average rate of 0.28/1000 athlete-exposures meaning that ACL tears occur at a rate almost half that of concussions. The most common mechanism of injury is noncontact in nature accounting for approximately 70-90% of all ACL ruptures (Boden et al., 2000; Krosshaug et al., 2007). Short-term complications of ACL injuries include muscular compensation, reduced return-to-activity, and recurrent injuries (Finsterbush et al., 1990; Fuentes et al., 2011; Lohmander et al., Roos 2004; Lustosa et al., 2011). Long-term implications include – but are not limited to – abnormal knee biomechanics, secondary injuries including meniscal lesions, and an increased risk of knee osteoarthritis (OA) (Butler et al., 2008; Fuentes et al., 2011; Lohmander et al., 2004; Neumann et al., 2008; Roos et al., 2004; Shimokochi and Shultz 2008). An estimated 45% of ACL injured individuals who undergo ACL reconstruction surgery develop early stages of knee OA within 10 years of the initial injury with a higher percentage of patients that used conservative treatments (non-surgical) developing early symptoms in the same time period (Lohmander et al., 2004; Roos et al., 2004). One of the most striking trends is the large gender disparity in injury rates, with females being at a 4-6-fold greater risk of ACL injury compared to their male matched-controls in the same sport (Hootman et al., 2007; Mihata et al., 2006). Therefore, research has aimed to characterize differential intrinsic (i.e., anatomy and knee geometry) and extrinsic (i.e., biomechanical and neuromuscular) factors responsible for ACL injury across genders. These factors are summarized in the schematic diagram in Figure 3 and will be further described in the sections below.

2.4 Biomechanical and neuromuscular mechanisms of acute ACL injuries

2.4.1 Biomechanical

Considering the anatomy and function of the ACL, it is evident that any event causing hyperextension of the knee with internal rotation of the tibia with respect to the femur or severe valgus motion may result in injury to the ACL among other structures (Markolf et al., 1995). Further mechanisms that have been proposed include: high shear forces on the knee joint from

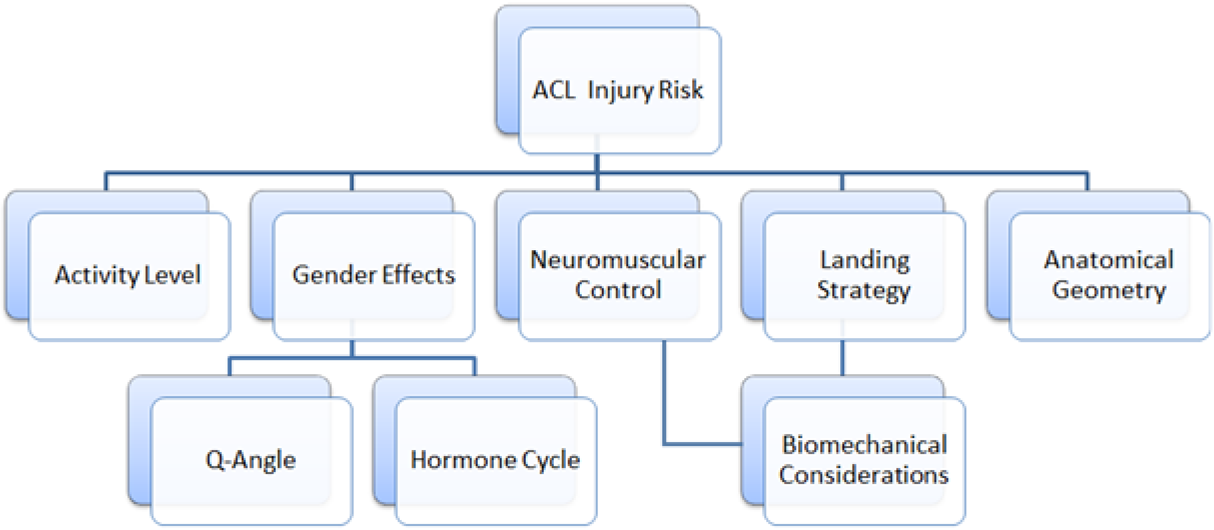

Figure 4: Factors associated with ACL injury.

quadriceps domination (Demorat et al., 2004); excessive anterior-posterior shear ground reaction force (GRF) (Fleming et al., 2001); internal rotation of the tibia with respect to the femur (Fleming et al., 2001); the "valgus collapse" mechanism (Hewett et al., 2005; Shin et al., 2009); and high axial compressive loads (Li et al., 1998; Yeow et al., 2011). Boden et al (2000) in a retrospective video analysis of 39 ACL injuries from the NBA and WNBA found that a combination of the above mentioned factors contributed to the majority of injuries. Specifically, an internally rotated tibia and extended knee in a deceleration or landing pattern were precursors to an eventual valgus collapse. Potentially injurious scenarios were shown *in vitro* by Markolf's group (1995) as well as more recently by Shin et al (2011) who demonstrated that combined loading scenarios involving anteriorly directed force on the tibia combined with an internal rotation moment at near extension, and anterior tibial force with valgus moment at angles greater than 10 degrees of sagittal flexion presented the largest amount of stress on the ACL.

Erect landing postures have been linked with increased ground reaction forces in jumping landing tasks as the increased stiffness throughout the lower limb translates in higher force generation (DeVita and Skelly 1992). Consequently, a conscious reminder for participants to land with a hyperflexed posture, especially at the hip and knee, results in significantly lower ground reaction forces and subsequently lower risk for injury (Blackburn and Padua 2009; Myers and Hawkins 2010). Secondly, knee valgus moment– or knee abduction moment – induced through poor neuromuscular coordination is one of the more notable biomechanical noncontact mechanisms of injury (Boden et al., 2000; Hewett et al., 2005). Valgus angle and ensuing loads placed at the knee joint during screening measures of drop landings highly predict future ACL injury risk as shown by a prospective study of female soccer players by Hewett and colleagues in 2005. Additionally, a large number of biomechanical mechanisms of ACL injury are dictated by both gender and subject-specific neuromuscular patterns that control kinematic posture during sport-related activities including natural mechanical axes in the lower limb, motor patterns, and kinematic landing variables. Understanding these neuromuscular patterns and how to train against these high-risk patterns is imperative to mitigate injury risk.

2.4.2 Neuromuscular

The identification of neuromuscular mechanisms of ACL injury has aimed to distinguish at-risk athletes from those at a lower risk of injury. Much of the work has been focused on identifying neuromuscular patterns that illicit biomechanically injurious outcomes and include: agonist-antagonistic muscle patterns, pre-motor times, and responses to external perturbations or disruptions in planned motor executions (Malinzak et al., 2001; McLean et al., 2010; McLean and Samorezov 2009; Sigward and Powers 2006; Wojtys et al., 2003).

Pre-planned motor predictions for carrying out tasks and reactive muscular activity to external perturbations are both important neuromuscular factors that affect how a movement is completed. The inherently random and unanticipated nature of sports often call for quick and unplanned motor events that an athlete must be able to adjust to without injury (Bessier et al., 2003; McLean et al., 2010; McLean and Samorezov 2009). During single leg landings, McLean et al (2010) found that pre-motor times in unanticipated reaction tasks were significantly related to the degree of knee abduction moment during the stance phase of the landing and suggested that increased time to activation could result in an increased risk of injury. In planned cutting tasks, participants tend to use a combination of co-contraction and selective motor responses to mitigate external coronal and transverse plane moments introduced in multi-planar movement (Bessier et al., 2003). Consequently in unanticipated cutting tasks, selective motor responses are abandoned for a generalized co-contraction response (Bessier et al., 2003). It has also been documented that these effects may be accentuated when the participant is fatigued (Borotikar et al., 2008; McLean and Samorezov 2009). Even though peak joint loads occur too quickly to be a direct result of voluntary muscular control, these studies stress that athletes may be able to adopt or “preprogram” safer involuntary movement patterns that may protect against acute knee ligament injury.

Adopting safe activity patterns in the form of neuromuscular training have been implemented into team training regimens as a proactive intervention to prevent ACL injuries. Several neuromuscular training intervention studies have proven to be effective in substantially reducing injuries in prospective cohorts of athletes, especially females (Mandelbaum, Silvers, Watanabe, Knarr et al., 2005; Myer, Ford, Palumbo and Hewett 2005; Myklebust et al., 2003; Hewett, Lindenfeld, Riccobene and Noyes 1999). Hewett et al (1999) looked at injury rates in a group of females in a standardized jump training protocol (adapted from Hewett et al 1996) compared to a control group of activity matched females. Both groups were additionally compared to a group of male controls. Results showed that untrained female athletes were 4.8 – 5.8 times more likely to sustain a knee injury than the control male group, and trained females were 1.3 – 2.4 times more likely to sustain an injury than male controls (Hewett et al., 1999). Despite the documented effectiveness of some training programs, ACL injuries still occur at an increasing rate. This trend is alarming and accentuates the need for alternative approaches to injury prevention.

2.5 'The gender phenomenon' - Gender disparities in injury risk

As mentioned previously, females are considered as being at a 4-6-fold increased risk of obtaining an injury to the ACL compared to their male counterparts in the same sport (Hootman et al., 2007; Mihata et al., 2006). Both intrinsic and extrinsic factors have been evaluated to understand the gender differences in injury risk.

2.5.1 Intrinsic factors

Intrinsic factors of injury are also known as 'unmodifiable' factors and cannot be changed under conscious control. Such factors include anthropometrics and joint geometry. Female hormone cycles and ambient temperature have also been proposed, however these are outside the scope of this thesis. Anthropometrics and the natural mechanical axis alignment in the lower limb (natural varus/valgus alignment) dictate loading patterns in the lower extremity during activities of daily living. Excessive varus/ valgus knee alignment may be one factor predisposing injury risk as Engin and Korde (1974) found up to 95% increases in joint contact forces *in vitro* for knees with 5 degrees of either varus or valgus alignment.

Joint geometry and morphology is an important component to the assessment of intrinsic ACL risk factors such as intercondylar notch width, shallow tibial plateaus, and steep tibial slopes (Hashemi et al., 2010; McLean et al., 2010). McLean et al (2010) demonstrated that intercondylar distance alone was not a strong predictor of high risk knee biomechanics, however the ratio of tibial plateau width-to- intercondylar distance was significantly related to an increased knee abduction moment in the stance phase of a single leg land and cut task. Tibial slopes have also been suggested to affect injury risk. The slope of the plateau is measured using medical imaging and is defined as the angulation of the plateau in reference to the horizontal (as in Hashemi et al., 2010). It has been found that increased posterior tilt of the tibial plateau is found in those patients who have sustained an ACL injury versus controlled healthy cases (Hashemi et al., 2010; Stijak, et al., 2008). An initial thought was the increased Q-angle could be a risk factor for injury as knee joint morphology is affected. However when adjusted for sex, there were no differences in trends found which suggests that joint geometry is not a sex-driven factor in injury risk (Hashemi et al., 2010). Further, Giffin et al (2004) revealed the effects of increasing posterior tilt of the tibial plateau via osteotomy on knee kinematics. They found that slope affected the resting position of the femur on the tibia (posterior slope caused a posterior resting position of the femur on the tibial plateau) with these shifts present throughout the full range of knee passive motion and concluded that an increased versus a decreased tibial slope may be protective to the ACL (Giffin et al., 2004). These findings elucidate the role of anatomical joint geometry in ACL injury risk and may be an important factor for prediction risk algorithms or injury prevention programs to identify athletes at high-risk of injury.

2.5.2 Extrinsic factors

2.5.2.1 Biomechanical

Gender-specific biomechanical differences during landing activities have been demonstrated. Hewett et al (2005) investigated the 'valgus collapse' mechanism and quantified its potential for distinguishing high-risk athletes of obtaining an ACL injury. In a prospective study of 205 adolescent female basketball, volleyball and soccer players, valgus positioning in the knee (i.e., frontal plane angles) was a positive indicator of ACL injury risk in the female during a bi-lateral drop vertical jump task (step off of a platform, land on both feet and immediately jump as high as possible in a vertical direction) (Hewett et al., 2005). The results showed that peak frontal plane angles of the knee during symmetrical bi-lateral landing were significantly greater in at-risk ACL injury females than in non-risk females. His team also found that there was significant correlation between peak GRF and peak coronal plane knee angle in prospective ACL injured females.

2.5.2.2 Neuromuscular

Gender differences in neuromuscular control of the lower limb during sporting activities have been identified and have illuminated a few of the ambiguities around the gender inequality in ACL injury risk. In general, Hewett et al (1996) showed that males exhibit a 3-fold increase in knee flexor (i.e., hamstring and gluteus activity) activation during landing compared to females. Work by Wojtys et al (2003) concluded that decreased levels of co-contraction of the quadriceps and hamstrings at the knee for females resulted in increased tibial rotation and increased ACL injury risk. This increased internal rotation of the tibia occurred in both a passive and active muscle state suggesting that these mechanics are likely to be present in highly dynamic athletic tasks (Wojtys et al., 2003). A decreased ratio of hamstrings-to- quadriceps activation has been shown in several investigations (Malinzak et al., 2001; Sigward and Powers 2006). These studies generally conclude that a co-contraction ratio of the hamstrings-to-quadriceps should be over 65% to be protective of the ACL in dynamic tasks, and that this criterion is typically observed more in male than female populations. Because of this, Cowling and Steele (2001) concluded that motor patterning in the male was more likely to lead to effective "muscle synchrony" (i.e., patterns that mitigate injurious micro-motions about the knee joint). Due to the above intrinsic and extrinsic evidence presented in this section, the study population for this thesis was chosen to include only female participants as they have an increased risk of ACL injuries in comparison to their male counterparts in the same sport.

2.6 Knee bracing in rehabilitation and prevention

2.6.1 Types of braces

The American Association for Orthopaedic Surgeons released a position paper in 1984 about the use of knee braces and classified three major types: 1) rehabilitative, 2) prophylactic, and 3) functional. Rehabilitative braces are temporarily worn following surgery (approximately 6-8 weeks). They typically cover from the mid-thigh through to the mid-calf and consist of an inner foam layer buttressed by long metal hinges laterally. They allow for swelling and are easy to take on and off to complete physiotherapy exercises post-surgery. Prophylactic knee braces (PKBs) are off-the-shelf type braces that are intended for use during activity to prevent injury, but are sometimes also worn to prevent further injury after minor injuries to knee structures. Finally, functional knee braces (FKBs) are typically custom-made by an orthotic company and are usually prescribed to be worn for ACL insufficiency including after ACL reconstruction. PKBs and FKBs are the two most widely used types of braces. A pictorial example of each type of brace is shown in Figure 4. Their efficacy for use in their respective domains has been tested both clinically and biomechanically with mixed results. In general, further research is needed to understand the relationship between brace use and biomechanical performance, namely, using a brace as an aid to prevent knee injuries and their potential effects on issues related to acute biomechanical and neuromuscular changes.

Figure 5: Three popular types of braces, the rehabilitation brace (left; Lenox Hill), prophylactic neoprene wrap (middle; DonJoy), and custom functional knee brace (right; OSSUR/CTi). A fourth type of brace (not pictured) is the patellofemoral brace designed to alleviate patellofemoral pain.

2.6.2 Early cadaveric and surrogate models for knee braces

Some of the earliest work on knee braces on their mechanical actions and effectiveness in mitigating injury were conducted on cadaveric specimens (Baker et al., 1989; France et al., 1987; Hofmann et al., 1984; Paulos et al., 1987; Paulos et al., 1991). Although criticized for not being representative of what occurs *in vivo* because of low physiological loads and reduced degrees of freedom, cadaveric studies were still able to generate considerable insights into the effectiveness of certain brace designs on protecting the knee during contact (Baker et al., 1989; Paulos et al., 1987; Paulos et al., 1991). Paulos et al (1987) used cadaveric modeling with instrumented prophylactic braces to identify design inadequacies including: MCL preload where brace design could place preloading on the MCL subjecting it to higher injury risk in contact; centre axis shift of the brace affecting the mechanical axis of the brace with the knee; premature joint line contact; and slipping of the brace during wear.

Surrogate modeling was another technique established to study the mechanical effects of the brace-knee composite on knee structure biomechanics. France et al (1987) used a combined *in vivo* and surrogate model study design to look at commercially available prophylactic lateral knee braces. NCAA Division I American football players were instrumented with the braces and lower limb EMG (vastus lateralis and vastus medialis) and asked to stand next to an Instron with a pendulum that induced lateral impact. Brace bending force and EMG were collected to assess proprioceptive feedback for anticipated and unanticipated lateral contacts. The same protocol was conducted on a metal surrogate bipedal skeletal model with supporting structures. The results of the surrogate model were found to be significantly associated with cadaveric tests (France et al., 1987), which supported the use of the surrogate method for assessing joint response to lateral loading. The main conclusion for the *in vivo* portion of the study was that even in ideal conditions where an impact could be anticipated, participants were unable to activate muscular elements quickly enough to offer considerable protection to the knee (France et al., 1987). The tests with the surrogate model showed that the braces were most effective in high mass/ lower velocity impacts with both the ankle and hip fixed and the knee at near extension and braces were least effective in low mass/ high velocity impacts with freely mobile ankle and hip joints and the near at 30 degrees of flexion (France et al., 1987). A secondary conclusion of this study and other similar modeling studies was that prophylactic braces at that time were biomechanically inadequate and recommended that braces should not be abandoned, but subjected to further design modifications and rigourous prospective testing (Hofmann et al., 1984; Paulos et al., 1991).

2.6.3 Bracing effectiveness through prospective studies

The efficacy of a knee brace during game play is important for the safety of the athlete, whether they are trying to prevent against a primary injury or cease secondary damage to an injured joint. Epidemiological studies comparing injury rates in braced versus nonbraced individuals are one of the easiest ways to examine how a brace performs in an athletic setting. Three divergent outcomes have been established with the examination of prophylactic bracing efficacy in athletic play: higher injury rates with the unbraced versus braced groups (Sitler et al., 1990), no difference in injury rates between the groups (Hewson et al., 1984), and higher injury rates in braced versus unbraced groups (Rovere et al., 1987). The nature of these epidemiological studies did not allow for insights into the mechanisms by which the knee braces tested influenced (or had no effect) on injury rates. Due to controversial conclusions and few investigations into brace effectiveness *in vivo* that utilized physiologically relevant forces and moments, systematic review studies have concluded that the known evidence for knee brace wear in the prevention of injury is controversial, and that rigorous biomechanical investigations on the effectiveness of knee brace use is needed before they are subsequently recommended for use in the prevention of injury (Pietrosimone et al., 2008; Rishraj et al., 2009).

Despite the debatable evidence for prophylactic knee braces in the prevention of injury, studies examining prophylactic bracing of the ankle in sports such as volleyball (where there is a high risk of ankle sprain) show promise for the idea of using bracing as a prevention technique. DiStephano et al (2008) used an 8-week intervention design investigating effect of lace-up ankle braces on approach and ground reaction forces generated during a jump vertical jump take-off. The bracing group showed overall decreased joint displacement during a drop landing and takeoff task yet no differences in vertical ground reaction forces were observed. This was achieved by compensatory action at the knee resulting in an increased knee flexion angle at ground contact which is typically a protective maneouvre for the lower limb (DiStefano et al., 2008). Pedowitz et al (2008) saw in an epidemiological study that prophylactic ankle bracing significantly decreased ankle injuries from an injury rate of 0.98/1000 exposures in the unbraced group to 0.07/1000 exposures in the braced comparison group over a time period of 7 years. Finally, Cordova and Ingersoll (2003) showed that the peroneus longus stretch reflex increased with brace application over an 8-week period bringing up the possibility of increased sensorimotor response with external ankle support. While the evidence presented on prophylactic bracing in the ankle joint is promising, one must be cognizant when attempting to generalize these results to the use of knee bracing in the prevention of injury. Each joint in the kinematic chain, although working in concert during dynamic activities, should be considered separately with the addition of a mechanical aid as its effect on proprioception and mechanical action may be different.

2.6.4 Bracing for ACL deficiency

The majority of research on bracing into the new millennium has been on the effects knee braces have on ACL deficient patients and whether these braces induce biomechanically-favourable changes that maximize performance without compromising function. ACL deficiency refers to those patients who are either a) absent of an ACL due to a previous rupture, b) have a malfunctioning ACL due to previous ligamentous injury, or c) are born with excess laxity during anterior tibial translation. The effects of knee braces have been tested biomechanically using kinematic and kinetic approaches, from a neuromuscular perspective using EMG amplitude and timing variables, and subjectively through the use of questionnaires on patients using a brace as a rehabilitative adjunct.

Brace studies on ACL deficient patients *in vivo* accompanied cadaveric and surrogate models of brace efficacy in the mid-to-late 1980's. It was generally found that braces may protect the knee during static or low energy impacts but have little effects in preventing anterior translation of the tibia when high loads or unanticipated movements are encountered (Cook et al., 1989). An analysis of cutting maneuvers and straight line running done by Cook et al (1989) saw an increase of cutting and running performance in the braced limb compared to the unbraced limb in ACL deficient athletes mostly due to subjective feelings of comfort to complete tasks with the brace. This effect was even greater for those patients that had only regained 80% of their quadriceps strength after injury (Cook et al., 1989). It is useful to note that the increased cutting performance was coupled with kinetics and kinematics that demonstrated an increase in shear forces in the braced limb which may place the deficient knee at a greater risk for re-injury as the mechanical action of the brace is in question for high loading scenarios such as sports-specific cutting tasks.

Brace wear has been demonstrated to affect muscle activity in the hamstrings (increase), quadriceps (decrease), and gastrocnemius (increase) (Lustosa et al., 2001; Ramsey et al., 2003). This may be a result of decreased knee flexion and concomitant decreases in axial and mediolateral knee motion with functional brace wear (Knutzen et al., 1987). Typically, the hamstring group is referred to as an ACL agonist as one of its main actions is to draw the tibia posteriorly with respect to the femur while alternatively the quadriceps are referred to as an ACL antagonist as they naturally allow for the tibia to draw anteriorly on the femur. Branch et al (1989) found that a brace had no significant effect on muscular timing during the stance and swing phase of the cut, but caused a significant decreases in peak EMG for the medial hamstrings and quadriceps in ACL deficient patients. It was concluded that one of two scenarios were possible with the given results: 1) the brace provided mechanical stability to the knee requiring less co-activation from the agonist/antagonist muscle groups; or 2) the brace alters joint position and therefore muscular patterns (Branch et al., 1989). More recent evidence suggests the latter to result to the former (Handular et al., unpublished; Ramsey et al., 2003). What is still unknown about bracing in ACL deficient patients is the effect of wearing a functional brace for rehabilitation and return-to-sport after a long period of use. It would be useful to note whether patients wearing a brace as a mechanical aid post-injury are at greater risk of re-injury due to biomechanical and neuromuscular changes.

Finally, brace wear in ACL deficient patients have led to subjective feelings of security prompting the patient to a potential early return-to-activity before the injured joint is ready to return to full workload (Birmingham et al., 2008). It is hypothesized that a brace may replace the loss of normal afferent responses which may be correlated to the increases subjective feelings of comfort with brace use after ACL injury (Cawley et al., 1991).

2.6.5 Bracing for healthy participants

Examining the effects of bracing on healthy individuals is a helpful tool to examine the unequivocal effects of brace application without the potential for confounding factors in ACL deficient patients such as time to injury, rehabilitation progress, brace wear, activity level, age, and sex. It has the potential to establish a baseline for the effects of bracing which can then be compared to application in affected populations. There is biomechanical evidence for altered mechanics and muscle activity with brace wear in healthy individuals which is not entirely surprising. A study by DeVita et al (1996) sought to determine if wearing a functional knee brace for walking and straight line running tasks caused healthy participants to adopt an ACL deficient gait. They found that hip and ankle torque increased while knee torque decreased which is characteristic of ACL deficient gait. This study demonstrated that wearing a functional brace during rehabilitation of the ACL may be one of the major causative factors in the adaptations of specific gait patterns for ACL deficient patients (DeVita et al., 1996). Secondly, in an unpublished study by Handular et al (submitted) it was found that the mechanical action of a custom made functional knee brace had no statistically significant effect on ACL strain between braced and unbraced trials on a cadaveric knee specimen with an instrumented ACL. It was concluded that the altered muscle forces in the braced conditions may have more of an effect on the protection of the knee than the mechanical action of the brace itself.

2.7 Literature review summary

The ACL is a commonly injured structure in the knee joint. Injury to the ACL is often a cause of excessive valgus motion of the knee paired with an extended limb posture and high ground reaction forces. The consequences of this injury are immense as it is the cause of a number of acute and chronic musculoskeletal deformities including reduced return to play, pain, loss of functional range of motion, and early onset knee osteoarthritis. Methods to prevent this injury from occurring are thus an important piece to the equation.

Prophylactic bracing is one of the two commonly adopted means of preventing an ACL injury. The evidence presented in the literature is still equivocal and epidemiological studies are inconclusive about the efficacy of the brace in protecting against injury (Sitler et al., 1990; Hewson et al., 1984; Rovere et al., 1987). Prophylactic braces have been tested on both ACL deficient and healthy participants. It has been documented that bracing tends to increase ACL agonists and decreases ACL antagonists with an overall effect to protect the ACL during movement (Ramsey et al., 2003). It is still debated whether the brace adds mechanical stability to the lower limb causing these neuromuscular changes, or whether the brace acts more like a proprioceptive feedback mechanism that offers little mechanical advantage but rather alters muscle activity to assist the ACL during dynamic activity (Branch et al., 1989; Handular et al., unpublished). Regardless of this chicken-or-egg paradox, neuromuscular changes have been demonstrated while wearing a brace for dynamic activity.

What is unknown and will be paramount to understanding the effects of brace wear worn over an acute period is the effect neuromuscular and biomechanical changes have over time after the removal of the brace. Specifically, it is unknown if the neuromuscular and biomechanical changes seen in wearing a brace for dynamic activities have residual effects that last after removing the brace. This is important information to gather as it may give insight into mechanisms that can be adopted for injury prevention after the removal of the brace, or alternatively methods for optimizing the benefits of brace wear without having to wear the brace for the main game event. In other words, if residual effects are present and they are protective of the knee, a brace could be worn for the warm-up period and removed for the game to have the benefits of the brace without having to be encumbered with its application during game play. **My thesis aims to fill a portion this knowledge gap, with the aim of characterizing residual differences in biomechanical and neuromuscular variables with prophylactic brace wear following an acute bout of standardized exercise as well as upon brace removal and 30-minutes post-exercise.**

Chapter 3: Evaluation of Acute Biomechanical and Neuromuscular Changes with Prophylactic Brace Wear

While prophylactic and functional bracing have been investigated for their role in knee ligament rehabilitation and injury prevention, major literature gaps remain. Namely, ideas are vague with respect to brace use and the biomechanical changes observed after acute doses of standardized exercise and the potential residual effects of these changes after the completion of exercise. This knowledge will have the potential to drive future research initiatives investigating the role of a knee brace after exercise, whether protective methods are necessary to reverse negative biomechanical and neuromuscular effects putting the knee at an increased risk upon brace removal or alternatively brace wear during the game could become an idea of the past. Additional details can be found in section 2.6 of the literature review chapter.

3.1 Purpose and hypotheses

The purpose of this thesis was to investigate the biomechanical and neuromuscular changes associated with acute brace wear and whether these changes had a residual effect lasting up to 30-minutes past the cessation of exercise. A cohort of healthy active female participants was recruited to complete a multi-session rigorous biomechanical analysis of limb kinematics, kinetics and electromyography (EMG) for unbraced (control) and braced (intervention) sessions. This investigation provided baseline data for future work in brace effectiveness. It was hypothesized that a main effect of brace would be present following exercise and remain 30-minutes of rest following brace removal that would be considered detrimental to the participant. The specific hypotheses tested were:

1) A prophylactic knee brace will have effects on biomechanical and neuromuscular variables before and following 30-minutes of standardized treadmill exercise. This hypothesis tested the main effect of the knee brace on biomechanical and neuromuscular factors. Specifically:

a. **Peak vertical ground reaction force (vGRF) and time to peak vGRF** will decrease in the braced versus unbraced trials (Rishiraj et al., 2012)

b. **Sagittal ankle, knee, and hip angles**, as well as **knee valgus angles at ground contact and peak vGRF**, will be *larger* in unbraced versus braced trials (i.e., more erect) (Hewett et al., 2005)

c. **Sagittal ankle, knee, and hip moments** will *decrease*, and **valgus knee moments** will be *larger* at the moment of **peak vGRF** in unbraced versus braced trials (Singer and Lamontagne 2008)

d. **EMG onset** in the **hamstrings** during landing will *increase* (i.e., have a delayed onset) with the braced condition (DeVita et al., 1996)

e. **EMG magnitude** in the **hamstrings** and **quadriceps** muscles will *decrease* with the braced condition (Handular et al., submitted)

2) Removing the brace will not have immediate effects on biomechanical and neuromuscular variables resulting from the braced exercise condition. This hypothesis aimed to identify immediate acute effects after brace removal. Specifically:

a. **Peak Fz and time to peak Fz** variables will *remain decreased* in the braced trials. Rishiraj et al (2012) demonstrated decreased peak Fz and time to peak Fz during drop landings after training sessions with a brace.

b. **Sagittal ankle, knee, and hip angles**, as well as **knee valgus angles at ground contact and peak vGRF**, will *remain larger* in the braced trials. An erect landing posture is often suggested to cause increased vGRF and an increased risk of injury (DeVita et al., 1992; Hewett et al., 2005). This conceptually *disagrees* with the predictions from hypothesis 2a, however more erect postures have been documented with brace wear. These two variables will have to be assessed in conjunction to examine the full effect of each variable to injury risk.

c. **Sagittal ankle, knee, and hip moments** will *remain decreased*, and **valgus knee moments** will *remain unchanged* at the moment of **peak vGRF** in the braced condition. Decreased moments are hypothesized to occur due to decreased sagittal plane angles and decreased peak vGRF values during landing.

d. **EMG onset** in the **hamstrings** during landing will *remain increased* in the braced condition. Implications for increased hamstrings onset with brace removal would suggest that the muscle may not be able to activate quickly enough to aid in the protection of the ACL during the landing task.

e. **EMG magnitude** in the **hamstrings** and **quadriceps** muscles will *remain decreased* in the braced trials. This suggests that the level of co-contraction will remain unchanged, just at a decreased absolute magnitude. The decreased magnitude of the hamstrings and quadriceps muscles may cause an increase in risk as the co-contraction response is suggested to be important in unanticipated tasks (Bessier et al., 2003).

3) Thirty-minutes of rest after the removal of the prophylactic knee brace <u>will not</u> have an <u>effect</u> on biomechanical and neuromuscular variables. This hypothesis aimed to test residual effects after delayed acute brace removal and supplement information on the link of brace wear with detraining. Specifically:

a. **Peak Fz and time to peak Fz** will *remain decreased* in the braced trials 30-minutes post-exercise

b. **Sagittal ankle, knee, and hip angles**, as well as **knee valgus angles at ground contact and peak vGRF**, will *remain larger* in the braced trials 30-minutes post-exercise

c. **Sagittal ankle, knee, and hip moments** will *remain decreased*, and **valgus knee moments** at the moment of **peak vGRF** will *remain unchanged* in the braced trials 30-minutes post-exercise

d. **EMG latencies** in the **hamstrings** during landing will *remain increased* in the braced trials 30-minutes post-exercise

e. **EMG magnitude** in the **hamstrings** and **quadriceps** muscles will *remain decreased* in the braced trials 30-minutes post-exercise

Figure 6 shows a graphical representation of the three hypotheses.

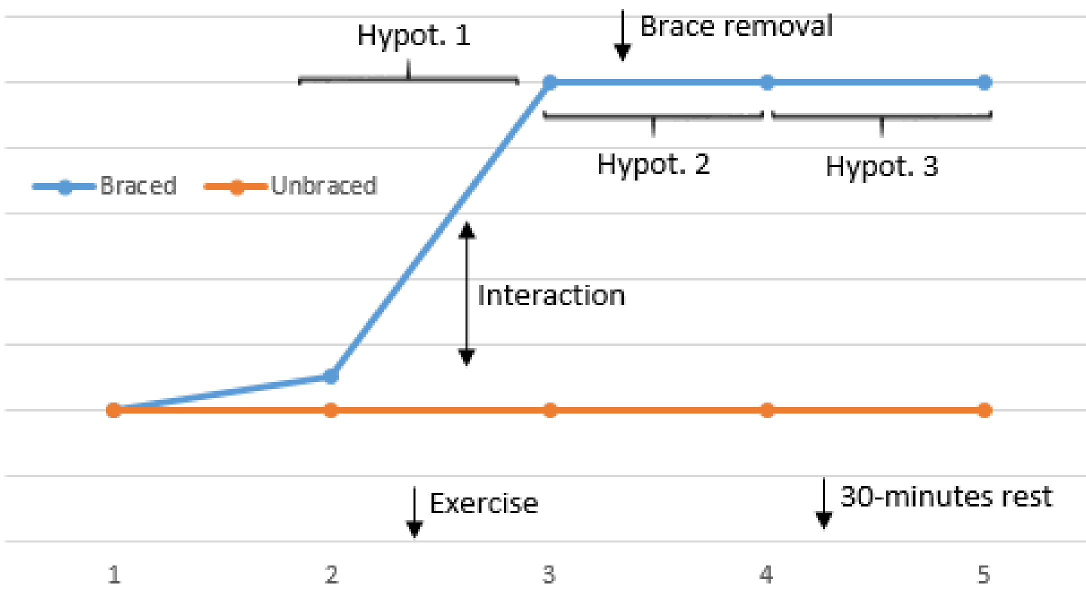

Figure 6: Graphical representation of the three hypotheses tested in this thesis.

3.2 Methods

3.2.1 Participants

Twelve female participants (age 21.4±2.1 years; height 1.69±0.04 m; body mass 63.0±7.0 kg) were recruited from the University population and were required to be active in a sport or rigorous activity at least once per week. Female participants were the representative population examined as they have an increased risk of ACL injury compared to their male counterparts (Hootman et al., 2007). Participants were excluded if they have a history of meniscus or anterior cruciate ligament injury, episodic "buckling" of the knee, any orthopaedic condition that would prevent them from participating in dynamic landing activities, or any other acute or chronic injury to knee structures within the past 18 months. A remuneration fee of $20 was provided to each participant for appreciation of their time. This study was approved through the University of Waterloo's Office of Research Ethics, and each participant provided informed consent.

3.2.2 Study design

A within-subjects control/intervention design was used for this study. The study consisted of two sessions per participant and each session was randomized. One session (unbraced) acted as a control session while the other session (braced) served as a comparison session during which the experimental protocol was repeated while the knee brace was worn immediately prior-to, during, and immediately post-exercise. The instrumentation and protocol were identical between sessions with the exception of the brace wear. Collection independent and dependent variables are summarized in Table 1.

The dependent variables in this dataset are important in determining injury risk in participants in both unbraced and braced conditions. Sagittal and frontal plane angles at force plate contact, peak vGRF, and the resulting ROM between these two times are an important indicator of landing posture during the impact event. Sagittal plane moments allow for the examination of the knee flexion/ hip extension paradox recognized as a new mechanism for potential ACL injury (Haskemi et al.., 2011). Frontal plane knee moment at peak vGRF will show valgus loading of the knee during peak impact. Normalized EMG in the hamstrings, quadriceps, and gastrocnemius allow for the calculation of co-contraction at the knee from these major muscle groups.

Table 1: Independent and dependent variables to be extracted from the dataset

Independent Variables	Dependent Variables
i. **Session** (unbraced vs. braced) ii. **Time** (5 levels; 2 pre-exercise and 3 post-exercise)	a) **Peak GRF** in the **vertical** direction, along with time to peak vGRF and the rate of loading b) **Sagittal** and **frontal** plane **ankle, knee and hip angles** at **force plate contact** c) **Sagittal** and **frontal plane ankle, knee, and hip angles** at **peak vGRF** d) **Sagittal** and **frontal** plane **ankle, knee, and hip** range of motion (**ROM**) from force plate contact to peak vGRF e) **Sagittal** plane **ankle, knee, and hip moments** at **peak vGRF** f) **Sagittal** plane **ankle, knee, and hip moment,** and **frontal** plane **knee moment** at **peak vGRF** g) Hamstrings, quadriceps, gluteus medius, and gastrocnemius **EMG activation/onset** **h) Co-Contraction ratios** of **quadriceps/hamstrings** and **quadriceps/gastrocnemius** during **preparatory phase** and **post-**

3.2.3 Experimental protocol

Participants were asked to come into the lab wearing spandex bottoms and a comfortable shirt. Upon giving informed consent, participants were pre-randomized into either session 1 (control) or session 2 (intervention). Next, each participant was fitted with a CT-I OTS prophylactic knee brace (OSSUR, Foothill Ranch, CA) on the right limb. The area where the brace covered the knee was traced on the participant with a washable marker to guide the placement of electrodes. This was repeated for the second session to allow for consistent placement of the electrodes across sessions.

3.2.3.1 Standardized measurement tasks

Instrumented participants were asked to complete 5 trials of a single-leg landing task at five time points during the collection. Each trial was be five seconds in length. The instructions to complete the task were as follows:

1) Single leg landings – participants were assessed during a single leg drop landing task involving stepping off of a 0.36 metre platform and landing with the right limb on the centre of the forceplate (Brazen et al., 2010). Each trial was five seconds, and participants were required to step off the platform, land, and achieve a state of balance without touching down with the contralateral foot in that time. Trials were discarded and re-collected if the participant could not achieve the three components of the landing in the allotted time.

3.2.3.2 Experimental protocol and conditions

The protocol was split into 5 time points, each outlined below:

1. Time 1 was a control condition where 5 single-leg drop landings were completed with no brace application.

2. Time 2 was a pre-exercise condition where the same 5 trials were completed. In the second session where the brace was involved, the brace was applied and used for the data collection in this section.

3. Next was the standardized exercise condition. The standardized exercise protocol involved a treadmill placed in the lab space. A treadmill protocol was chosen to simulate an applicable situation where a prophylactic brace may be used by an active

population. The protocol was based off the Standard Bruce Treadmill Protocol which is a standardized VO^2 submaximal treadmill test (Bruce 1971). Participants wore a Polar Heart Rate monitor with an adjustable chest strap to measure heart rate during the protocol. Resting heart rate, rating of perceived exertion (RPE) using the 20-point Borg scale and a visual analog scale for rating of perceived comfort (RPC) for the braced limb were obtained before the commencement of the test. The warm-up lasted a total of 3-minutes at a grade of 10% and speed of 1.7 mph. Heart rate, RPE, and RPC were taken at the end of the warm-up as dictated by the protocol. The participant had their work rate increased based on the protocol in Table 1 and were required to remain at a work rate that elicits a heart rate of approximately 65-80% HRmax (calculated as 0.65 – 0.8 X [220 – age]). The work rate was increased systematically every three minutes until this HR range was reached. Heart rate and RPE were recorded at every stage (every 3 minutes) and were subsequently recorded every 3 minutes once a stable work rate was reached to ensure that all variables remain stable and were within the proper bounds of the protocol (HR = 65-80% of HRmax; RPE = 13-16 out of 20, which have been shown to correlate with heart rate). The total protocol lasted 28-minutes including the warm-up, stage progression, and cool-down. The cool down was administered with the treadmill set to the same work rate as the warm-up stage. In the second session, the above practice was repeated with the exception of the brace applied and used for the duration of the exercise protocol.

Table 2: Progression for the Standard Bruce Treadmill Protocol (Bruce 1971)

Stage	Minutes	%Grade	Speed (km/h)	Speed (mph)	METS
1	3	10	2.7	1.7	4.6
2	3	12	4.0	2.5	7.0
3	3	14	5.4	3.4	10.2
4	3	16	6.7	4.2	12.9
5	3	18	8.0	5.0	15.0
6	3	20	8.8	5.5	16.9
7	3	22	9.6	6.0	19.1

4. Time 3 was a post-exercise condition where the 5 single-leg landing trials were completed within 5 minutes of completing the exercise protocol. In the second session, the brace was worn during these trials.

5. Time 4 was another post-exercise condition where the 5 single-leg landing trials were completed with the brace **taken off** in the intervention session to examine the immediate effects of brace removal.

6. Time 5 was a 30-minutes post-exercise condition to assess residual biomechanical effects of the brace post-exercise. In this time, the participant sat and rested for a total period of 30-minutes. The same 5 single-leg landing trials were completed for a total of 25 trials over the span of the session. For session two, the brace was not worn for the completion of these trials.

After session two, a subjective questionnaire directed toward feelings of brace comfort, brace support, prophylactic efficacy, and overall remarks was administered to each participant to match potential subjective feelings with biomechanical outcomes (see Appendix F).

3.2.4 Experimental set-up

Instrumentation associated with EMG, kinematic and kinetic data collection were employed. Specifically, bi-lateral skin areas overlying the muscle bellies of the biceps femoris, rectus femoris, gluteus medius, and medial gastrocnemius were shaved with a single-use razor and scrubbed with isopropyl alcohol wipes. Sixteen Ag-AgCl blue sensor electrodes (two per muscle with 2.5 cm interelectrode spacing) were placed on the muscles of interest in parallel to the muscle fibres. Maximum voluntary exertions were completed for each muscle for normalization (procedures are summarized in Appendix B). Next, eight rigid marker clusters (each equipped with 4 Optotrak smart markers in a rectangular orientation) on the left and right foot, left and right shank, left and right thigh, lower back and upper back were affixed to the body using double-sided tape and Velcro straps. From these clusters, forty-two bony landmarks were digitized in the calibration trial on the lower limbs and trunk. The probe was used to locate the following landmarks in relation to the rigid body marker clusters: bi-lateral first metatarsal, first toe, fifth metatarsal, superior midfoot, lateral midfoot, heel, lateral and medial tibial condyles, tibial tuberosity, lateral and medial femoral condyles, greater trochanter anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), and acromion; as well as the ventral sacrum, spinous processes of the T10 and C7, suprasternal notch, and xiphoid process. Palpation and landmarking techniques are summarized in Appendix C.

Kinetic data was collected with a single AMTI OR-7-2000 forceplate (AMTI, Waterdown, MA) imbedded in the floor at a rate of 1600 Hz and collected synchronously with EMG data [CMRR 115 dB @ 60 Hz, bandpassed from 10-1000 Hz, input impedence = 10 GΩ]. Kinematic data was collected with a 12-camera Optotrak Certus Motion Capture system (Northern Digital Inc., Waterloo, ON) at a frame rate of 80 Hz. The global coordinate system was placed at one corner of the forceplate with +Y pointing up, +X pointing anteriorly and +Z pointing to the left (see Appendix D for a schematic diagram of the collection space). Sampling rates were chosen based off of pilot work in a previous investigation. Specifically, Fast Fourier Transform of the Fz channel of the forceplate data in single leg drop landings show the majority of the signal contained frequencies less than 100 Hz. Additionally, the signal bandwidth of surface EMG is 10 – 500 Hz. Therefore the sampling rate of 1600 Hz for the forceplate and EMG is a conservative oversample of the signal that follows well within the restrictions of Nyquist

3.2.5 Data analysis

All data was treated with custom Matlab software (Mathworks, Inc., Natick, MA) and Visual 3D software (C-Motion, Inc., Kingston, ON). Kinematic data was filtered with a dual pass 2nd order Butterworth filter at a cutoff frequency of 15 Hz (based on pilot results looking at the vertical position of the first metatarsal marker). Moment and joint angle data was calculated in Visual3D software using a custom built pipeline. EMG magnitude data from times 1, 2, 3, 4 and 5 was zeroed (e.g., mean removed), full wave rectified and linear enveloped using a single pass 2nd order Butterworth filter to mimic muscle twitch response and electromechanical delay. The cutoff frequency used was 3 Hz and processing was done in

Matlab software (Mathworks, Inc., Natick, MA). Force plate data was treated with a dual pass 2nd order Butterworth filter with a cutoff frequency of 70 Hz and used to characterize the following force-related variables: peak Fz, time to peak Fz, and rate of loading. This cutoff was chosen based off pilot results for the vertical force channel during drop landing trials and balances the technique of removing signal noise while not attenuating peak force. A combination of Fast Fourier Transform and Residual Analyses were used to determine cutoff frequency. Moment data was run in a separate pipeline with the kinematics and force data both filtered with a 2nd order low pass Butterworth filter at a cutoff frequency of 15 Hz (Bisseling and Hof 2006). Data on the effect of filter cutoff for moment data variables are not included in this thesis but are available upon request.

A number of dependent variables were extracted from the dataset and are summarized in Table 1. Repeated data with multiple trials were averaged into a single value. Peak Fz, time to peak vertical ground reaction force (vGRF) from initial ground contact (IGC) (defined as when the vertical force trace reached 10 Newtons as used in Brazen et al., 2010) was expressed in the Fz force plate channel. Additionally, impulse (defined as a rate of loading) from initial ground contact to peak vGRF was calculated based off the following equation:

$$\int_{t1}^{t2} F\,dt$$ [1]

All joint angles and moments (sagittal and frontal) were calculated in Visual3D software. All raw EMG data was synchronized with force plate data to assess activation and latencies with respect to initial ground contact with the force plate. Preparatory phase EMG activation was extracted with the 0th time point being defined at IGC (negative values indicate activation before

[1] F = peak vGRF (in Newtons); t2 = time of peak force (ms); t1 = time of ground contact (ms)

force plate contact). Processed EMG was normalized to %MVC in custom Matlab software to investigate EMG magnitude during drop landings. Normalized data was then used to calculate the time-varying co-contraction index for the rectus femoris/biceps femoris and the rectus femoris/gastrocnemius muscle pairings. The equation used (equation 2) was adapted from Hubley-Kozy et al (2009). Since a time-varying integral with magnitude was wanted instead of a percentage value, the adaptation shown was chosen over the equation used in Hubley-Kozy et al (2009) which incorporated a multiplication of (1/100).

$$\sum_{i=1}^{N} \left[\frac{Lower\ EMG_i}{Higher\ EMG_i}\ x\ (Lower\ EMG_i +\ Higher\ EMG_i) \right] \quad [2]$$

Minimum co-contraction values as well as the time of the minimum value were calculated as a representation of the time of least stability during both the preparatory phase (165 ms before force plate contact to initial force plate contact) and the post-landing phase (from force plate contact to the moment of maximum knee flexion during the landing).

3.2.6 Statistical analysis

3.2.6.1 Preliminary analysis

The hypotheses were tested with a series of descriptive and inferential statistical tests. First, a paired t-test was used to determine inter-session differences between the control measures. If no significance was found, then time 1 measures were left out of the subsequent analysis of variance (ANOVA) models for the collected data. As an initial overview, a single two-factor ANOVA (brace and time) from time points 2-5 was run on the dataset. However, the

[2] Adapted from Hubley-Kozy et al., (2009). Note that their equation added a multiplication by (1/100) to obtain a percentage whereas our equation gives a simple ratio that incorporates EMG magnitude.

next step involved three separate analyses to test each of the three hypotheses specifically. The hypotheses were each tested with two factor repeated measures ANOVAs with time and session as the independent variables. Time points 2 and 3 which represent the time points immediately prior to and after exercise, respectively were used in the analysis of the first hypothesis, time points 3 and 4 were used for the second hypothesis (brace removal), and finally time points 4 and 5 were used for testing the third hypothesis (after 30-minutes of rest). In the event of an interaction effect, separate paired t-tests were run to test the effect of session on variable differences. All statistical tests were run in SPSS software (SPSS Version 19.0, SPSS Inc., Chicago, IL, USA) using an alpha value of 0.05.

3.2.6.2 Secondary analysis

Since a small sample of healthy participants were used, additional statistical measures were employed to test for the possibility of statistical insignificance due to large variability in one of either the braced and unbraced conditions. Standard deviation values for all variables were tested with two-way repeated measures ANOVAs for each of the three hypotheses. Specifically, we were interested in determining whether variance was affected by brace condition (i.e. would wearing a brace result in more consistent within-participant responses across the 5 repeated trials at each time point). Additionally, measures of clinical equivalency were completed to test for mean similarities (Barker et al., 2001; 2002). This approach complements the initial difference testing approaches (ANOVA) and addresses the issue of lower-than-desired power. Four variables were chosen for this analysis and are considered important variables in the assessment of ACL injury risk. These are: frontal plane knee angle at force plate contact (Hewett et al., 2005), hamstring onset (DeVita et al., 1996), sagittal plane range of motion in the knee joint from force plate contact to the moment of peak vGRF (kinematic assessment of landing stiffness), and rate of loading/ impulse values (kinetic assessment of stiffness). Equivalency of means testing was done by assessing the confidence interval of the mean difference between unbraced and braced values at each time point. Mean differences were then represented as a percentage of the unbraced value at each time point. A confidence interval of 90% was used for the investigation (Barker et al., 2001). Mean differences that fell outside the ±10% bounds of the reference value were considered as clinically different (Barker et al., 2002). Mean differences and CI that fell within the ±10% bounds were considered clinically the same. Mean differences that fell within the ±10% bounds but had CI that extended beyond the bounds had a significance that was unclear.

Chapter 4: Results

4.1 Analysis of variance (ANOVA) results – Means

Variables at time 1 are not included in the analysis as paired t-tests for time 1 indicated no significant differences between sessions except for one variable [sagittal ankle angle at peak Fz (p=0.021); included in Table 3]. As an initial overview, a single two factor ANOVA was performed on the dataset from times 2-5 to look at significant main effects and post-hoc results across all time points. This analysis was not included within the original hypotheses, however it allowed for a basic analysis of whether significant differences are present somewhere across time points.

Only significant main effects of brace wear and interaction effects of Time*Brace will be reported in this thesis. Tables 3 and 4 display overall ANOVA results (specifically, F and p statistics for main effects, interaction effects, and post-hoc tests). No significant effects of brace were observed for any force, joint angle, or joint moment variable. A significant effect of brace was observed for rectus femoris/gastrocnemius co-contraction during the post-landing phase (p=0.030). Interaction effects were present for only two variables: frontal plane ankle ROM from force plate contact to peak vGRF (p=0.001) and frontal knee ROM from the same two events (p=0.007). Despite this relatively limited support for brace-related effects on dependent variables in this initial test, ANOVAs specific to each hypotheses were run and are presented in greater detail in the sections below.

4.1.1 Force Variables: Peak Fz, Time to Peak, and Rate of Loading

Table 5 contain F and p statistics for the two-way ANOVAs performed on the force variables. Results indicated no significant main effect of brace on all force variables, and no significant interaction effects between independent variables (brace and time). This trend was present for each of the three hypotheses tested. Figures of differences in means with standard deviations for force variables are included in Appendix I (Figure I-1).

4.1.2 Joint angles

Tables 6 and 7 contain F statistic and P-value results from the two-way ANOVAs for hypotheses one, two and three for joint angle variables at initial ground contact, at peak Fz, and the range of motion (ROM) of the ankle, knee, and hip joint between those events. Additionally, included in Appendix J is the exported Visual3D reports of sagittal ankle, knee, and hip range from force plate contact to maximum knee flexion across all participants. Main effects of brace were observed for frontal plane ankle (p=0.010) and frontal plane hip (p=0.033) ROM for hypothesis one (pre- and post-exercise). Specifically, frontal ankle ROM was larger in the unbraced condition than the braced condition (12.3 ± 5.6^{o} vs. 8.9 ± 4.0^{o} respectively) and frontal hip ROM was smaller in the unbraced condition than the braced condition (1.4 ± 1.4^{o} vs. 3.4 ± 3.1^{o} respectively). Additionally, interaction effects of brace and time were observed for the frontal plane ankle ROM (p=0.030) and frontal plane knee ROM (p=0.003) in hypothesis one.

Table 3: F and p statistics and Post-Hoc results for the two-way ANOVA run across time points 2-5. Paired t-tests for time 1 are also presented.

	Variable	F			p			Post Hoc						T-Test Time 1
		Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*	*Time 2 vs Time 3*	*Time 2 vs Time 4*	*Time 2 vs Time 5*	*Time 3 vs Time 4*	*Time 3 vs Time 5*	*Time 4 vs Time 5*	*p*
Force Variables	Peak Fz	0.198	6.460	1.22	0.665	**0.001***	0.312	0.005	**0.338**	0.338	**0.011***	**0.004***	0.071	0.238
	Time2Peak	0.353	2.192	0.856	0.564	0.108	0.473	0.256	0.793	0.466	**0.003***	**0.006***	0.217	0.101
	RoL	0.022	2.647	1.086	0.885	0.114	0.355	0.405	0.688	0.256	**0.006***	**0.002***	**0.01***	0.160
Angle - Initial Ground Contact	Sagittal Ankle	0.819	3.851	0.926	0.387	**0.044***	0.407	0.368	**0.049***	0.069	**0.039***	**0.033***	0.784	0.108
	Frontal Ankle	0.964	0.525	1.620	0.349	0.668	0.230	0.413	0.249	0.506	0.600	0.955	0.694	0.643
	Sagittal Knee	0.187	0.969	1.158	0.675	0.420	0.342	0.441	0.518	0.183	0.957	0.230	0.205	0.089
	Frontal Knee	0.097	0.810	1.269	0.762	0.498	0.303	0.474	0.839	0.314	0.419	0.326	0.053	0.946
	Sagittal Hip	0.375	0.230	2.141	0.563	0.875	0.131	0.440	0.965	0.698	0.190	0.982	0.556	0.579
	Frontal Hip	0.548	0.733	1.639	0.487	0.501	0.242	0.312	0.156	0.581	0.541	0.906	0.470	0.090
Angle - Peak Force	Sagittal Ankle	2.411	0.145	0.388	0.152	0.932	0.762	0.622	0.93	0.905	0.448	0.497	0.879	**0.021***
	Frontal Ankle	0.014	2.029	0.752	0.908	0.150	0.437	0.684	0.132	0.207	0.055	0.126	0.925	0.835
	Sagittal Knee	0.555	2.041	1.527	0.475	0.158	0.243	0.152	0.442	0.825	0.085	**0.029***	**0.033***	0.096
	Frontal Knee	2.106	1.534	0.265	0.181	0.228	0.850	0.191	0.548	0.768	0.308	0.072	0.097	0.511
	Sagittal Hip	0.611	0.978	1.989	0.464	0.425	0.152	0.650	0.306	0.667	0.463	0.326	0.075	0.314
	Frontal Hip	2.381	1.834	1.900	0.174	0.207	0.181	0.095	0.319	0.518	**0.006***	0.152	0.805	0.101
Angle - ROM	Sagittal Ankle	0.038	2.710	0.621	0.850	0.112	0.540	0.537	0.112	0.144	**0.030***	**0.023***	0.770	0.435
	Frontal Ankle	1.157	1.842	7.594	0.307	0.161	**0.001***	0.111	0.629	0.879	**0.025***	**0.004***	0.265	0.490
	Sagittal Knee	0.395	3.288	0.691	0.544	**0.034***	0.565	0.189	0.562	0.170	0.331	**0.019***	**0.011***	0.962
	Frontal Knee	2.218	1.440	4.963	0.171	0.253	**0.007***	0.116	0.805	0.181	0.078	0.495	0.506	0.742
	Sagittal Hip	0.154	1.318	1.280	0.708	0.299	0.311	0.185	0.172	0.950	0.474	0.240	0.379	0.113
	Frontal Hip	4.146	3.534	1.135	0.088	**0.036***	0.362	0.054	0.102	0.602	0.391	**0.022***	0.092	0.579

<u>Table 4:</u> F and p statistics, as well as post-hoc results and paired t-tests for Time 1 (control) for each variable.

	Variable	F			p			Post Hoc						T-Test Time 1
		Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*	*Time 2 vs Time 3*	*Time 2 vs Time 4*	*Time 2 vs Time 5*	*Time 3 vs Time 4*	*Time 3 vs Time 5*	*Time 4 vs Time 5*	*p*
Moment - Peak Force	Sagittal Ankle	0.021	1.949	0.228	0.889	0.143	0.876	0.727	0.102	0.052	0.163	0.223	0.759	0.365
	Sagittal Knee	0.252	3.788	0.946	0.629	0.063	0.405	**0.032***	0.175	0.220	**<0.001***	0.057	0.823	0.711
	Frontal Knee	2.136	1.209	1.076	0.182	0.328	0.378	0.324	0.700	0.077	0.461	0.511	0.197	0.338
	Sagittal Hip	0.020	4.990	0.609	0.892	**0.031***	0.513	0.050	0.098	0.176	0.072	0.080	0.911	0.659
EMG Onset	RF Onset	0.674	0.797	0.787	0.429	0.504	0.510	0.546	0.507	0.544	0.798	0.209	0.232	0.100
	BF Onset	0.676	0.718	1.461	0.428	0.548	0.243	0.370	0.223	0.581	0.395	0.975	0.410	0.829
	G Onset	0.000	0.454	1.889	0.999	0.716	0.151	0.697	0.669	0.563	0.408	0.367	0.689	0.743
	GM Onset	0.131	0.944	0.274	0.725	0.388	0.713	0.318	0.552	0.900	0.151	0.026	0.235	0.235
CoContraction Index	Prep Phase RF/BF	0.391	2.378	0.270	0.547	0.135	0.736	**0.028***	0.191	0.552	0.973	0.064	**0.042***	0.482
	Prep Phase RF/G	2.447	4.210	1.095	0.152	**0.035***	0.353	0.071	**0.038***	0.506	0.519	**0.046***	**0.011***	0.928
	Post Phase RF/BF	0.126	0.448	0.529	0.731	0.721	0.666	0.796	0.844	0.421	0.215	0.231	0.521	0.150
	Post Phase RF/G	6.645	3.090	2.665	**0.030***	**0.044***	0.068	0.244	0.499	0.060	0.267	**0.037***	**0.034***	0.444

Table 5: F and p statistics for force variables Peak Fz, time to peak Fz, and rate of loading (RoL). Results are displayed for Hypothesis 1 (A), Hyothesis 2 (B), and Hypothesis 3 (C).

A.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Peak Fz	0.515	12.308	0.457	0.488	**0.005***	0.513
Time2Peak	0.049	1.433	0.723	0.829	0.256	0..413
RoL	0.05	0.75	1.215	0.827	0.405	0.294

B.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Peak Fz	0.744	9.247	0.043	0.398	**0.011***	0.839
Time2Peak	0.325	14.958	0.023	0.580	**0.003***	0.882
RoL	0.098	11.229	0.405	0.760	**0.006***	0.538

C.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Peak Fz	0.031	3.985	4.699	0.864	0.071	0.053
Time2Peak	0.937	1.718	1.039	0.354	0.217	0.330
RoL	0.562	9.789	1.731	0.469	**0.010***	0.215

Concerning the interaction effects, paired t-tests revealed a significant effect of brace for time 2 (brace application; p=0.024) and no significance for time 3 (post-exercise; p=0.797) for frontal plane ankle ROM. For frontal knee ROM, t-tests showed the opposite trend with non-significance for time 2 (p=0.374) and significance for time 3 (p=0.043).

For hypothesis two, no significant effects of brace were observed for the joint angle variables. An interaction effect of brace and time were observed for the frontal plane knee ROM (p=0.012). Paired t-tests revealed a significant effect for time 3 (p=0.017) and a non-significant effect for time 4 (brace removal; p=0.237). All other statistical results from the second hypothesis are summarized in Table 3b.

For hypothesis three (30-minutes rest), no main effect of brace was observed. Sagittal plane knee angle at peak Fz had a significant interaction effect of brace and time ($p=0.042$). Paired t-tests showed no significant of session for time 4 ($p=0.501$) as well as time 5 ($p=0.203$). Mean differences and standard deviations for the frontal ankle, knee and hip variables for all three hypotheses are displayed in Appendix I (Figures I-2 – I-7).

4.1.3 Joint moments

Results (F and p statistics) of the two-way ANOVAs for hypothesis one, two and three of the joint moments at peak vertical ground reaction force (vGRF) are included in Table 8. As well, Appendix K contains Visual3D exports for overall sagittal joint moment excursion ranges from initial force plate contact to maximum knee flexion for unbraced and braced conditions for all five time points. Hypothesis one saw no significant main effect of brace for sagittal and frontal plane moments at the ankle, knee, and hip. As well, no interaction effects of brace and time were observed.

The second hypothesis had no main effects of brace for all variables tested. Significant interaction effects of brace and time were observed for frontal knee moment ($p=0.020$). Sagittal knee moments increased from 11.4±26.3 Nm (UB) and 7.1±31.4 Nm (BR) to 19.9±31.8 Nm (UB) and 12.7±26.3 Nm (BR) upon brace removal. This represents a 74.6% and 77.7% increase for unbraced and braced conditions respectively. Paired t-tests for frontal knee moment revealed a non-significant effect of session for time 3 ($p=0.097$) and time 4 ($p=0.597$).

Table 6: ANOVA F and p statistics for the first (A) and second (B) hypotheses. Significant main effects of brace, time, and any interaction effects are highlighted with an asterisk (*) and bolded.

A.

	Variable	F			p		
		Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Initial Ground Contact	Sagittal Ankle	0.326	1.977	1.609	0.579	0.187	0.231
	Frontal Ankle	0.930	0.729	2.279	0.358	0.413	0.162
	Sagittal Knee	0.169	1.426	1.650	0.689	0.258	0.225
	Frontal Knee	0.087	1.171	2.955	0.773	0.302	0.114
	Sagittal Hip	1.110	0.035	0.062	0.320	0.855	0..810
	Frontal Hip	0.470	0.003	7.466	0.510	0.960	0.023
Peak Force	Sagittal Ankle	2.588	0.258	0.192	0.139	0.622	0.671
	Frontal Ankle	0.000	0.176	1.439	0.992	0.684	0.126
	Sagittal Knee	0.191	1.136	0.023	0.671	0.309	0.881
	Frontal Knee	3.459	0.046	0.052	0.090	0.834	0.824
	Sagittal Hip	1.114	0.019	0.539	0.322	0.893	0.484
	Frontal Hip	1.787	1.958	0.178	0.218	0.199	0.685
ROM - Initial Ground Contact to Peak Force	Sagittal Ankle	0.089	0.409	0.458	0.771	0.537	0.514
	Frontal Ankle	10.104	3.049	6.345	**0.010***	0.111	**0.030***
	Sagittal Knee	0.111	0.411	0.006	0.745	0.535	0.942
	Frontal Knee	3.476	6.605	14.736	0.089	**0.026***	**0.003***
	Sagittal Hip	0.477	1.045	0.278	0.509	0.336	0.612
	Frontal Hip	6.132	7.118	3.636	**0.033***	**0.024***	0.086

B.

	Variable	F			p		
		Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Initial Ground Contact	Sagittal Ankle	0.484	5.653	0.331	0.503	**0.039***	0.578
	Frontal Ankle	1.220	0.293	1.797	0.295	0.600	0.210
	Sagittal Knee	0.245	0.128	1.509	0.630	0.727	0.245
	Frontal Knee	0.170	0.534	1.573	0.688	0.480	0.236
	Sagittal Hip	0.806	2.935	0.519	0.395	0.125	0.492
	Frontal Hip	0.073	0.866	0.015	0.794	0.379	0.905
Peak Force	Sagittal Ankle	1.485	0.623	2.510	0.251	0.448	0.144
	Frontal Ankle	0.064	4.697	1.088	0.806	0.055	0.321
	Sagittal Knee	0.314	5.279	0.555	0.588	**0.044***	0.473
	Frontal Knee	0.894	0.996	0.001	0.367	0.342	0.982
	Sagittal Hip	1.419	0.808	0.141	0.268	0.395	0.717
	Frontal Hip	0.398	12.687	2.595	0.546	**0.007***	0.146
ROM - Initial Ground Contact to Peak Force	Sagittal Ankle	0.119	6.425	0.000	0.737	**0.030***	0.992
	Frontal Ankle	0.004	6.949	0.437	0.95	**0.025***	0.524
	Sagittal Knee	0.025	1.525	0.018	0.877	0.243	0.896
	Frontal Knee	4.236	4.767	9.456	0.067	0.054	**0.012***
	Sagittal Hip	0.238	0.455	0.433	0.639	0.519	0.529
	Frontal Hip	4.093	0.740	0.761	0.078	0.415	0.408

Table 7: ANOVA F and p statistics for the third hypothesis accounting for thirty minutes of rest post-exercise. Significant main effects of brace, time, and any interaction effects are highlighted with an asterisk (*) and bolded.

Variable		F			p		
		Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Initial Ground Contact	Sagittal Ankle	0.180	0.079	0.147	0.680	0.784	0.710
	Frontal Ankle	0.963	0.164	0.597	0.349	0.694	0.597
	Sagittal Knee	0.452	1.835	0.576	0.517	0.205	0.465
	Frontal Knee	0.049	4.814	2.101	0.829	0.053	0.178
	Sagittal Hip	0.082	0.002	2.585	0.783	0.970	0.152
	Frontal Hip	0.704	0.211	0.782	0.429	0.660	0.406
Peak Force	Sagittal Ankle	1.968	0.024	0.766	0.191	0.879	0.402
	Frontal Ankle	0.035	0.009	0.484	0.855	0.925	0.503
	Sagittal Knee	0.706	6.364	5.643	0.423	**0.033***	**0.042***
	Frontal Knee	1.682	3.430	0.335	0.227	0.097	0.577
	Sagittal Hip	0.687	4.632	4.557	0.439	0.075	0.077
	Frontal Hip	0.622	0.067	0.411	0.460	0.805	0.545
ROM - Initial Ground Contact to Peak Force	Sagittal Ankle	1.016	0.090	0.575	0.337	0.770	0.466
	Frontal Ankle	0.510	1.394	0.783	0.491	0.265	0.397
	Sagittal Knee	0.543	9.652	1.287	0.478	0.011	0.283
	Frontal Knee	1.635	0.480	0.132	0.233	0.506	0.725
	Sagittal Hip	0.553	0.909	0.874	0.481	0.372	0.381
	Frontal Hip	2.454	4.018	0.559	0.168	0.092	0.483

No main effects of brace were observed for the third hypothesis (effects after 30-minutes of rest). One significant interaction effect of brace and time was observed for sagittal knee moment (p=0.032). The paired t-test showed a non-significant effect of session for both time 4 (p=0.554) and time 5 (p=0.773). Mean differences and standard deviations for all moment variables for all three hypotheses are displayed in Appendix I (Figure I-8 – I-9).

Table 8: F and p statistics for hypothesis one (A), hypothesis two (B), and hypothesis three (C) for sagittal plane ankle, knee, and hip, as well as frontal plane knee moments at peak force.

A.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Sagittal Ankle	0.284	0.129	0.069	0.606	0.727	0.799
Sagittal Knee	0.114	5.865	1.027	0.743	**0.036***	0.335
Frontal Knee	3.015	3.167	0.304	0.113	0.106	0.593
Sagittal Hip	0.019	10.260	0.984	0.895	**0.015***	0.354

B.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Sagittal Ankle	0.012	2.268	0.744	0.913	0.163	0.408
Sagittal Knee	0.245	44.158	0.274	0.633	**<0.001***	0.613
Frontal Knee	1.826	0.161	7.884	0.210	0.698	**0.020***
Sagittal Hip	0.547	4.181	0.004	0.483	0.080	0.953

C.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Sagittal Ankle	0.060	0.100	0.075	0.812	0.759	0.790
Sagittal Knee	0.290	0.054	6.742	0.605	0.823	**0.032***
Frontal Knee	1.047	1.977	1.872	0.336	0.197	0.208
Sagittal Hip	0.185	0.014	1.484	0.685	0.911	0.278

4.1.4 Electromyography

4.1.4.1 Muscle onset

Table 9 contains F and p statistics for all muscle onset dependent variables. No main effect of brace or time was observed for all variables throughout all three hypotheses. Additionally, no interaction effects of brace and time were observed.

4.1.4.2 Co-Contraction Index (CCI)

Analysis of variance results for co-contraction index (CCI) dependent variables are presented in Table 10. For the first hypothesis, no main effect of brace was observed for all variables. No interaction effects were observed as well.

For the second hypothesis, a main effect of brace was observed for CCI between rectus femoris and gastrocnemius muscles in the post-force plate contact phase (p=0.015). CCI values were larger for the braced condition in comparison to the unbraced condition [18.80±12.21 (UB) versus 33.33±23.86 (BR); 77.3% increase]. No interaction effects were present.

Table 9: F and p statistics for EMG onset for all three hypotheses (A, B, and C respectively). No main effects of brace or time were observed for these dependent variables.

A.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
RF Onset	1.803	0.388	0.896	0.206	0.546	0.364
BF Onset	0.031	0.873	0.007	0.863	0.370	0.934
G Onset	2.549	0.159	0.000	0.139	0.697	1.000
GM Onset	0.360	1.103	0.039	0.562	0.318	0.848

B.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
RF Onset	0.794	0.069	1.926	0.392	0.798	0.193
BF Onset	0.695	0.784	2.410	0.422	0.395	0.149
G Onset	0.005	0.740	2.191	0.947	0.408	0.167
GM Onset	0.001	2.417	1.035	0.975	0.151	0.333

C.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
RF Onset	0.000	1.602	0.059	0.991	0.232	0.813
BF Onset	1.567	0.733	0.006	0.237	0.410	0.940
G Onset	0.706	0.169	0.052	0.419	0.689	0.824
GM Onset	0.134	0.155	0.831	0.722	0.701	0.382

RF = rectus femoris, BF = biceps femoris, G = gastrocnemius, GM = gluteus medius

The third hypothesis had a main effect of brace for CCI between the rectus femoris and gastrocnemius in the post-force plate contact phase (p=0.047). CCI values were larger for the braced condition (average of 18.84±10.77) in comparison to the unbraced condition (average of 26.65±15.52). Braced values where therefore 41.5% larger in comparison to unbraced values.

Table 10: F and p statistics for co-contraction index (CCI) variables. A. Results for the first hypothesis. B. Results for the second hypothesis. C. Results for the third hypothesis.

A.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Pre CC Min RF/BF	1.169	11.590	2.389	0.303	**0.006***	0.150
Pre CC Min RF/G	2.177	3.966	0.391	0.168	0.072	0.391
Post CC Min RF/BF	0.040	0.164	2.536	0.846	0.693	0.140
Post CC Min RF/G	3.924	2.063	1.274	0.073	0.179	0.283

B.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Pre CC Min RF/BF	0.405	0.001	0.050	0.539	0.972	0.828
Pre CC Min RF/G	2.108	0.475	0.559	0.177	0.507	0.472
Post CC Min RF/BF	0.122	2,433	0.932	0.734	0.150	0.357
Post CC Min RF/G	8.558	1.474	0.629	**0.015***	0.253	0.446

C.

Variable	F			p		
	Brace	*Time*	*Brace*Time*	*Brace*	*Time*	*Brace*Time*
Pre CC Min RF/BF	0.301	5.584	0.113	0.596	**0.042***	0.744
Pre CC Min RF/G	1.567	10.079	1.465	0.242	**0.011***	0.257
Post CC Min RF/BF	0.074	0.445	1.192	0.791	0.521	0.303
Post CC Min RF/G	5.293	6.221	5.630	**0.047***	**0.034***	**0.042***

Pre CC Min RF/BF = Minimum CCI value between rectus femoris and biceps femoris muscles in the preparatory phase
Pre CC Min RF/G = Minimum CCI value between rectus femoris and gastrocnemius muscles in the preparatory phase
Post CC Min RF/BF = Minimum CCI value between rectus femoris and biceps femoris muscles from force plate contact to maximum sagittal knee flexion
Post CC Min RF/G = Minimum CCI value between rectus femoris and gastrocnemius muscles from force plate contact to maximum sagittal knee flexion

Additionally, an interaction effect of brace and time was observed for the CCI value between rectus femoris and gastrocnemius muscles during the post- force plate contact phase (p=0.042). Paired t-tests revealed a significance of session for time 4 (p=0.030) but not for time 5 (p=0.411). Main effects of brace and the main effects of time for all EMG variables for all three hypotheses are displayed in Appendix J (Figure J-10 – J-13).

4.2 Analysis of variance (ANOVA) results – Standard deviations

Only a main effect of brace was investigated for the analysis of variance involving standard deviations for a cohort of variables. The variables chosen were: peak force, time to peak force, and impulse; sagittal plane ankle, knee, and hip angles at initial ground contact, peak force, and the ROM of those joints between those two times; frontal plane knee angle at initial ground contact and peak force; Sagittal plane ankle, knee, and hip, as well as frontal plane knee moments at peak force; muscle onset; and co-contraction index variables. Table 8 includes a summary of major results

4.2.1 Force

No main effect of brace was observed for across-trial standard deviation force plate variables for all three hypotheses tested. None of the variables were approaching significance with the lowest p-value observed for peak Fz force during hypothesis 3 (p=0.266).

4.2.2 Joint angles

No main effects were observed for across-trial standard deviations for all sagittal and frontal plane joint angles (ankle, knee, and hip joints) during initial ground contact, peak force, and the range of motion of those joints between ground contact and peak force. Sagittal plane hip angle at initial ground contact for the third hypothesis, though insignificant, was approaching significance at an alpha level of 0.05 (p=0.072).

4.2.3 Joint moments

No significant main effect of brace for across-trial standard deviation values was observed for any sagittal and frontal plane moment variable (at the ankle, knee, and hip joints) for all three hypotheses tested.

4.2.4 Electromyography

4.2.4.1 Muscle onset

Only one variable (gastrocnemius onset) had a main effect of brace for variable standard deviations for the third hypothesis (p=0.034). Variance values were larger for the braced condition in comparison to the unbraced condition (19.27±15.84 ms (UB) versus 29.36±22.16 ms (BR); 52.4% increase). All other variables (quadriceps onset, hamstrings onset, and gluteus medius onset) did not approach significance.

4.2.4.2 Co-Contraction Index (CCI)

Co-Contraction Index (CCI) variables had a main effect of brace for across-trial standard deviations at several instances. For the first hypothesis, CCI value post force plate contact between the rectus femoris and gastrocnemius muscles had a significant main effect of brace (p=0.014) Average variance values were 5.34±3.39 in the unbraced condition compared to 10.19±7.41 in the braced condition, making braced variance values 90.8% higher than unbraced values. For the second hypothesis, brace effects were observed for rectus femoris/biceps femoris and rectus femoris/gastrocnemius CCI variables post force plate contact (p=0.047 and p=0.026

respectively). Braced variances were larger for the rectus femoris/ biceps femoris CCI variable compared to unbraced variances [1.82±1.78 (UB) versus 2.71±2.64 (BR); 48.9% increase]. Similarly, braced variances were larger for the rectus femoris/gastrocnemius CCI variable [3.76±2.61 for unbraced versus 10.63±9.83 for the braced condition; 182.7% increase in the braced condition]. Finally for the third hypothesis, CCI value for rectus femoris/gastrocnemius prior to force plate contact in the preparatory phase saw a main effect of brace ($p=0.046$). Following trends presented above, braced variances were higher (3.50±2.87) in comparison to unbraced variances (1.93±1.37). This is an increase of variance by 81.3% for the braced condition.

Table 11: F and p statistics for the variances of chosen dependent variables.

	Variable	Hypothesis 1		Hypothesis 2		Hypothesis 3	
		F	p	F	p	F	p
Force	Peak Fz	1.354	0.269	1.288	0.281	1.372	0.266
	Time2Peak	0.429	0.526	0.189	0.672	0.817	0.385
	RoL	0.012	0.913	0.068	0.799	0.202	0.662
Initial Ground Contact	Sagittal Ankle Angle	0.540	0.478	0.035	0.854	0.019	0.893
	Sagittal Knee Angle	0.173	0.685	0.059	0.812	1.753	0.212
	Frontal Knee Angle	0.443	0.519	0.795	0.392	0.436	0.523
	Sagittal Hip Angle	0.507	0.491	0.114	0.743	3.954	0.072
Peak Force	Sagittal Ankle Angle	0.023	0.881	0.013	0.911	3.213	0.101
	Sagittal Knee Angle	1.236	0.290	1.896	0.196	0.238	0.635
	Frontal Knee Angle	0.458	0.513	1.855	0.200	2.304	0.157
	Sagittal Hip Angle	0.316	0.585	0.337	0.574	2.043	0.181
ROM	Sagittal Ankle Angle	0.149	0.707	0.008	0.932	0.160	0.697
	Sagittal Knee Angle	0.810	0.387	3.073	0.107	1.832	0.203
	Sagittal Hip Angle	2.699	0.129	2.129	0.172	1.008	0.337
Peak Force - Absolute	Sagittal Ankle Moment	0.444	0.519	0.000	0.980	0.032	0.861
	Sagittal Knee Moment	1.456	0.253	0.535	0.480	0.201	0.663
	Frontal Knee Moment	1.295	0.279	4.355	0.061	3.455	0.090
	Sagittal Hip Moment	0.075	0.790	0.091	0.768	0.947	0.351
Muscle Onset	Rectus Femoris Onset	0.150	0.706	0.097	0.761	0.039	0.848
	Biceps Femoris Onset	0.063	0.806	0.034	0.858	0.049	0.829
	Gastrocnemius Onset	1.076	0.322	1.769	0.210	5.889	**0.034***
	Gluteus Medius Onset	0.166	0.692	0.001	0.971	0.174	0.685
Co-Contraction Index (CCI)	Preparatory Phase RF/BF	0.013	0.912	0.332	0.577	0.042	0.841
	Preparatory Phase RF/G	1.773	0.210	2.528	0.143	5.334	**0.046***
	Post-Contact RF/BF	0.627	0.445	5.147	**0.047***	1.714	0.223
	Post-Contact RF/G	8.542	**0.014***	6.841	**0.026***	3.897	0.080

4.3 Equivalence testing

Testing for mean similarities using confidence intervals was completed for four variables: frontal plane knee angle at force plate contact, hamstring onset, sagittal plane range of motion in the knee joint from force plate contact to the moment of force plate contact, and rate of loading. All results are in reference to the ±10% bounds of the unbraced condition for each time point.

Table 12: Mean difference and confidence intervals for frontal plane knee angle at force plate contact.

	10% CI	MEAN DIFF.	90% CI
TIME 1	-1.55	-0.06	1.44
TIME 2	-1.83	-0.35	1.13
TIME 3	-1.38	0.95	3.28
TIME 4	-2.34	0.11	2.56
TIME 5	-2.02	0.58	3.19

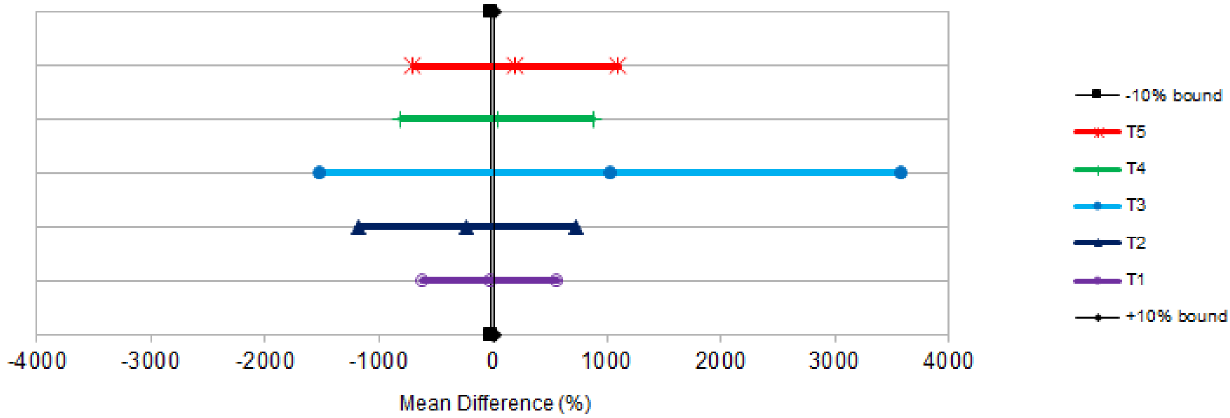

Figure 7: Confidence interval of the mean difference between the unbraced and braced trials for frontal plane knee angle at force plate contact at each time point. ±10% was the threshold considered to indicate clinically equivalent values.

90% confidence intervals for mean differences for frontal plane knee angle at peak force fall outside the ±10% reference bounds at all five time points (Figure 7). Table 9 shows that mean differences fell below the -10% bound for times 1 and 2, and above the +10% bound for times 3, 4, and 5.

Mean differences and 90% confidence intervals for hamstring onset have means that both fall within and fall without the ±10% reference bounds. Mean differences for times 1, 2, and 3 fall within the ±10% bounds but have CI that extend beyond the bounds indicating unclear interpretation of equivalence while times 4 and 5 fall to the left of the -10% bound meaning the means are not equivalent. Results are displayed in Figure 8 and Table 10.

Table 13: Mean differences and confidence intervals for hamstring onset across all time points.

	10% CI	MEAN DIFF.	90% CI
TIME 1	-12.24	1.72	15.68
TIME 2	-11.35	0.72	12.78
TIME 3	-11.44	1.35	14.14
TIME 4	-35.27	-14.71	5.86
TIME 5	-44.22	-15.72	12.77

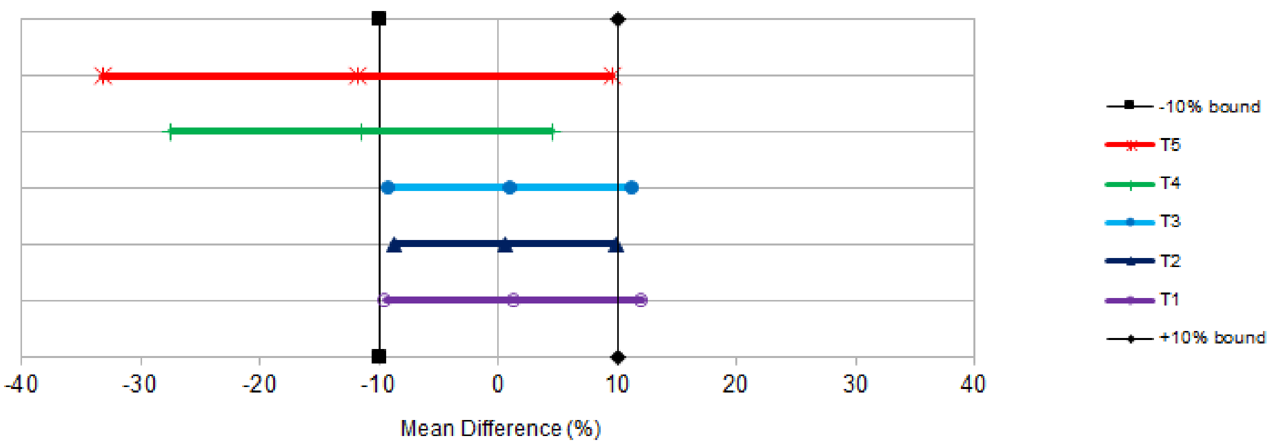

Figure 8: 90% Confidence intervals of the mean differences for hamstrings onset.

Table 14: Mean differences and 90% confidence intervals for time 1-5 for sagittal plane knee ROM.

	10% CI	MEAN DIFF.	90% CI
TIME 1	-2.19	0.06	2.32
TIME 2	-2.49	0.48	3.44
TIME 3	-2.32	-0.39	1.53
TIME 4	-1.92	0.22	2.36
TIME 5	-4.05	-1.53	1.00

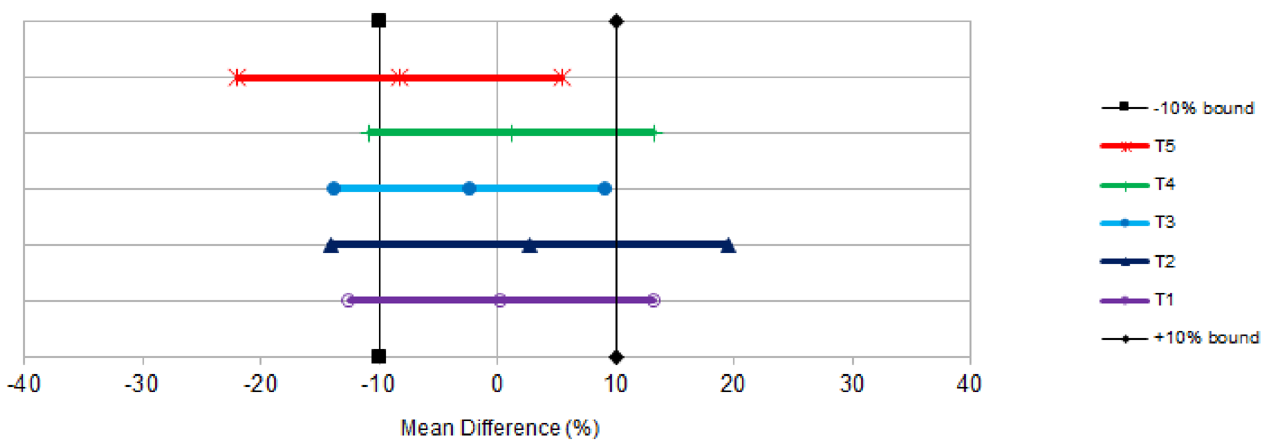

Figure 9: Mean differences and 90% confidence intervals for sagittal plane knee ROM.

Mean differences for sagittal plane knee ROM were contained within the ±10% bounds for times 1-5 with confidence intervals extending beyond the bounds indicating unclear equivalence of means. Percentage values and mean differences are shown in Table 14 and Figure 9 respectively.

Mean differences and confidence intervals for rate of loading fell left of the -10% bound for times 1 and 2 (non-equivalence), and fell within the ±10% bounds for the remainder of the times with CI beyond the bounds (unclear equivalence). Results are presented in Table 12 and Figure 10.

Table 15: Mean differences and confidence intervals for rate of loading for times 1-5.

	10% CI	MEAN DIFF.	90% CI
TIME 1	-12115	-5532	1052
TIME 2	-13337	-3556	6226
TIME 3	-5399	1712	8823
TIME 4	-4895	371	5638
TIME 5	-956	3255	3255

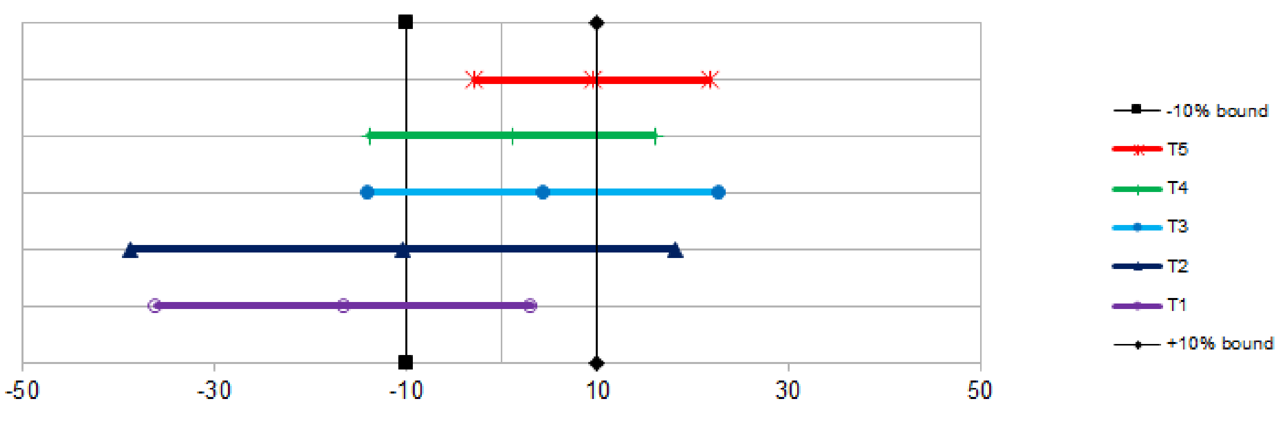

Figure 10: Mean differences and confidence intervals for rate of loading.

Table 16: Overall equivalence results for frontal plane knee angle, hamstrings onset, knee ROM, and rate of loading. The + sign represents mean differences that were larger than the +10% bound and the - sign represent mean differences that were below the -10% bound. The UNC sign represents mean differences that were unclear.

	TIME 1	**TIME 2**	**TIME 3**	**TIME 4**	**TIME 5**
Frontal knee angle at force plate contact	-	-	+	+	+
Hamstrings onset	UNC	UNC	UNC	-	-
Sagittal plane knee ROM	UNC	UNC	UNC	UNC	UNC
Rate of loading	-	-	UNC	UNC	UNC

Interpretation of all results are presented in the discussion section below.

Chapter 5: Discussion

The purpose of this thesis was to examine biomechanical and neuromuscular changes with acute brace wear before and following standardized repetitive exercise. There were three main hypotheses, each characterizing neuromuscular and biomechanical changes at a different point through the protocol: hypothesis one looked at changes pre- and post-exercise with expected changes and interaction effects for joint angles, joint moments, muscle onset, and co-contraction; hypothesis two examined changes upon brace removal with the expected changes from hypothesis one to remain changes (continued main effect of brace with no interaction effect); and hypothesis three characterized changes after 30-minutes of rest after the cessation of exercise with an expected main effect of brace with no interaction effect (same as expected in hypothesis 2). There was a lack of statistical significance for a main effect of brace for the majority of vGRF variables, joint angles, joint moments, and electromyography variables when using an analysis of variance testing the protocol overall and again when testing each of the three hypotheses using separate ANOVAs. This suggests that the initial hypotheses set out in this thesis were not satisfied. Analysis of variance tests on variable standard deviations (variances) showed a lack of significance as well, suggesting that wearing a brace did not affect the variance at each variable. Finally, measures of equivalency for four variables (frontal plane knee angle at force plate contact, hamstring onset, sagittal plane knee range of motion, and rate of loading) revealed that equivalence in means for times 1, 2 and 3 for hamstrings onset, all of knee ROM means, and times 3, 4 and 5 for rate of loading variables. These results suggest that prophylactic brace wear for healthy participants may have minimal effects on biomechanical and neuromuscular variables following standardized treadmill exercise. The equivalence tests further suggest that with the exception of the sagittal knee ROM variables, a main effect of time may be the larger determinant in changes in mean differences and not the brace, further suggesting that the effect of the brace within the bounds tested in this thesis is minimal.

The lack of significance for the main effect of brace for joint angles during single leg landing is surprising. Evidence from Singer and Lamontagne (2008) demonstrated that wearing a shell brace during gait for healthy participants resulted in significantly less knee flexion during the swing phase which the authors proposed was due to increased stiffness that the brace imparts on the knee joint. Ramsey et al (2003) also demonstrated in a small sample case analysis of four male ACL-deficient patients that brace wear during a maximum single-legged horizontal jump caused decreased knee flexion angles during landing in two of four participants. This evidence had suggested that brace wear during a different motion such as single leg landing may have an effect on joint angles during the landing phase and a impart a potential effect on overall landing stiffness. No changes between unbraced and braced landing sessions in the current investigation is therefore difficult to interpret. It may be due to the nature of the single leg landing movement as a highly dynamic loading scenario and the inability of the brace to generate enough resistance on the knee joint to cause changes in knee flexion angle during the landing phase. Additionally, work by Hewett et al (2005) identifying frontal plane knee angle at ground contact as a risk factor for ACL injury could make one postulate that since knee braces have rigid medial and lateral constraints, the brace could have an effect on frontal knee motion and therefore an effect on injury risk. Equivalency tests for this variable demonstrated that means were not the same and therefore could be considered clinically significant. To elaborate, mean differences at time 1 were more varus (knee adduction) than unbraced angles for time 1 (baseline), and time 2 mean differences increased varus angle suggesting that brace application caused a slightly more varus knee alignment which is considered a less risky knee positioning with respect to ACL injury risk. Mean differences for times 3, 4, and 5 show a valgus (knee abduction) knee positioning in comparison to baseline values. This suggest that the intervention of the exercise caused average knee positioning to switch to a more valgus positioned knee during ground contact. This test also demonstrates that although there were non-equivalence in means, the differences were likely driven by the intervention of exercise and less so by the application of the knee brace.

Results revealed a few variables experiencing an interaction effect of brace and time meaning that changes are dependent on an interaction of both variables. An interaction effect for frontal plane ankle ROM for the first hypothesis is tough to interpret as it suggests a difference in ankle eversion/inversion between unbraced and braced conditions dependent on exercise. Since ankle frontal plane analyses are not generally looked at when assessing ACL injury risk and therefore will not be interpreted further in this thesis. Interaction effects for frontal plane knee ROM following brace removal (hypothesis 2) suggest that frontal plane motion in the braced knee is dependent on the removal of the brace following exercise, with changes observed after exercise with the braced conditions (*more* ROM) returning to unbraced values (*less* ROM) immediately as the brace is removed. This result suggests the brace seems to somehow cause excess frontal plane motion that is eradicated after brace removal. Though this result does not seem expected, it is notable to mention that the brace does not seem to have a residual effect on frontal knee motion after brace removal. The final joint angle interaction effect was sagittal knee angle at peak force at hypothesis 3 (30-minutes of rest) with non-significant session difference at either time 4 or time 5. The non-significant sessional differences are surprising and potentially due to large variances. Looking at the mean differences (Appendix I), trends indicate that after the braced session, participants land with increased knee flexion at peak force after 30-minutes of rest while after the unbraced session there is little difference. This increased knee flexion may be an attempt to re-establish pre-exercise knee flexion values as the exercise intervention (though not significant) caused a decrease in knee flexion at peak force (i.e., more erect). This may further add to the interaction effects seen with frontal knee ROM in that statistically significant residual effects of brace wear seemed to not remain upon brace removal and 30-minutes of rest for joint angle and joint moment variables.

Increased stiffness in the lower limb during landing – or more specifically decreased joint range of motion during landing - has been shown to increase peak vertical ground reaction forces (vGRF) (DeVita et al., 1992; Fong et al., 2011; Laughlin et al., 2010; Myers and Hawkins 2010). Some authors have suggested that a more erect landing posture may lead to an increased risk of ACL injury due to the increased risk of internal rotation and valgus collapse with an extended knee position (Fleming et al., 2000; Markolf et al., 1995). Further, Laughlin et al (2011) used kinematic and kinetic inputs from 15 female participants for a 3-Dimensional lower limb model and found that during soft landings versus stiff landings there was an 11% decrease in the peak ACL force suggesting that increased joint flexion during landing may be protective of the ACL. The current study found no significant differences in sagittal plane angles coupled with no significant changes in peak vGRF variables between braced and unbraced sessions showing that landing stiffness did not change between sessions. This finding disagrees with the findings of Rishiraj et al (2012) who found that peak vGRF values were significantly smaller for braced versus unbraced conditions. They also found that peak vGRF values systematically decreased as time with the brace worn increased suggesting an acclimatization period with prolonged brace wear respect to loading (Rishiraj et al., 2012). Their study used 23 males from collegiate level sports (basketball and field hockey) and fitted each with a custom fit functional knee brace. Differences between the Rishiraj study and this thesis may explain the differences in results observed. First, a larger sample of males was used instead of a smaller sample of females. Second, the custom fit brace may have had a factor, and third the length of time wearing the brace was much higher in the Rishiraj study which may have allowed for acclimatization to the brace over a longer period of time. Decreased knee flexion in braced conditions shown by other authors suggest a potentially more injurious joint position, however decreased ground reaction forces would suggest a decrease in the loading experienced by the lower limb. It is further possible that the brace itself may offer some alternate distribution of forces that are not measureable within the bounds of this thesis.

Equivalency tests for sagittal knee ROM and rate of loading reiterated the results from the ANOVA tests in that changes in landing stiffness remained relatively unchanged between unbraced and braced conditions, as well as between time points in the protocol. First, differences between unbraced and braced means at all time points for sagittal knee ROM were all within the ±10% bounds and were considered unclear with respect to equivalency. Therefore even on a clinical level, landing stiffness defined by lower limb kinematics was not considered different between unbraced and braced session at all five time points throughout the experimental protocol. Second, mean differences for rate of loading were not equivalent for time 1 and 2 (less than the 10% bound) meaning that mean differences for loading at these two times were less than the unbraced reference condition. This could be accounted for by either lower peak Fz values or an increased time to peak force value. From a kinetic point of view this suggests that during these two time points, participants were landing in a less stiff manner. Mean differences for times 3, 4, and 5 were all equivalent meaning no clinical significance in means. Comparing landing stiffness between kinematic and kinetic approaches, overall participants increased their landing stiffness as the protocol progressed dictated by the less stiff landing at the beginning of the protocol for the kinetic approach. Similarly to the frontal plane knee angle at ground contact, it seems likely that the changes observed are a result of the exercise intervention than the brace intervention suggesting that the brace had little effect on the single leg landing maneouvre tested in this thesis.

A lack of an effect for brace for muscle onsets for the rectus femoris, biceps femoris, gastrocnemius, and gluteus medius suggests that the prophylactic brace did not significantly alter neuromuscular onset of the muscles between sessions. Additionally, no interaction effects show that onset is not dependent on brace nor time point in the session. This result is not expected as it is becoming more known in the literature that knee braces typically cause changes in muscle firing instead of offering enough mechanical restraint to protect the knee itself (Handular et al., *submitted*; McNair et al., 1996). Two of the muscles of real importance in the assessment of knee motion and injury risk are the rectus femoris and the biceps femoris, as those two muscles offer protection at the knee by working antagonistically to one another. Equivalency tests for hamstring onset showed that mean differences were equivalent for times 1, 2, and 3 of the experimental protocol and were non-equivalent and less than the -10% bound for times 4 and 5. This proposes that following exercise and after the removal of the brace, hamstring onset occurs sooner in comparison to the unbraced time one baseline value. This result is difficult to interpret as it seems for this variable that changes due to the brace do not occur until after the cessation of exercise and after the brace is *removed*. Unlike the previous scenarios where the effect of time was likely a stronger determinant of outcome compared to the effect of brace, it seems as though in the case of hamstring onset that the effect of the brace (or brace removal) had a stronger effect on the outcome variable. Perhaps the muscle is compensating for a loss of proprioception that the brace may have provided however further research would be required in order to test this hypothesis. With an increased cocontraction present in the absence in a change in sagittal knee joint position, it can be hypothesized that compensation in the form of increased joint compression was used once the brace was removed. In order to test the hypothesis of a loss of proprioceptive feedback, a joint replication task with the use of a biodex could be employed (as in McNair et al., 1996). Finally, the one interaction effect present for cocontraction of the rectus femoris/gastrocnemius for the third hypothesis during the post-contact phase demonstrated that braced values had a significantly higher CCI value than unbraced values after exercise which declined back to unbraced values after 30-minutes of rest. This suggests that the brace may have had an effect causing in increase in cocontraction between these two muscles and in turn potentially creating more joint stability or joint compression while not changing joint ROM. Considering that this effect occurred once the brace was removed with little evidence of a similar trend during the wearing of the brace, it seems wise to hypothesize that, like the hamstring onset, that the muscles were potentially compensating for the loss of the brace sensation once removed. Work by Baltaci et al (2011) showed that proprioception is affected with brace wear using a visual feedback program to guide the participant through a joint position task. Again, further research would be required to demonstrate this effect within the protocol tested in this thesis.

There is potential for a prophylactic knee brace to have a different effect on those indicated as high-risk individuals for ACL injury. A sub-analysis was completed on the current dataset to assess if any individuals would be considered at a higher risk than the others and additionally how the brace affected these participants during the intervention session. The variables were assessed from the control unbraced condition and included: frontal knee angle at initial ground contact, normalized and absolute pVGRF, sagittal plane knee ROM from initial ground contact to pVGRF, muscle onset, and cocontraction. It is useful to note that no one participant had injurious tendencies in all categories. Therefore some subjectivity had to be used to ultimately determine the one or two persons likely to be at higher risk for injury. Two participants were chosen to have a potential increased risk of injury: participant 5 (P5) and participant 7 (P7). Based on one variable alone, P5 was considered a high-risk participant as their frontal knee angle at initial contact had the greatest degree of valgus positioning during the single-leg landings throughout the 5 time points. Hewett et al (2005) have demonstrated that frontal plane knee positioning may be a large factor in injury risk. Secondly, P7 was chosen as they demonstrated the lowest degree of sagittal plane ROM during landing throughout the 5 time points as well as delayed gluteus medius onset and decreased cocontraction following exercise. These EMG findings suggest that, especially after the exercise session, the participant had altered hip neuromechanics and decreased knee cocontraction which may affect injury risk. For P5, the knee brace was associated with a more varus knee positioning which is thought to be more protective of the knee. Additionally, the brace was associated with increased pVGRF, slightly decreased sagittal joint ROM from initial contact to pVGRF, and slightly increased cocontraction in the pre-landing phase. These results suggest a potential for decreased risk due to the changed knee positioning in the frontal plane alone, however the other results are more difficult to interpret as they actually suggest an increased risk (i.e., more erect posture with higher pVGRF). For P7, the knee brace had the following effects: small decrease in pVGRF, increase in sagittal joint ROM from initial ground contact to pVGRF, increase in varus knee positioning, slightly delayed rectus femoris and biceps femoris onset, quicker gluteus medius onset, and slightly decreased cocontraction between rectus femoris and gastrocnemius. These results suggest a small neuromuscular effect by a changing joint positioning and in turn affecting muscular activity. Overall, the brace may have a larger effect on certain people or populations but whether the effect is greater in an at-risk population requires further investigation.

The findings presented in this work have potential positive implications regarding brace wear and ACL injury risk. As the effects of longitudinal brace wear are still elusive in the literature, this investigation was the first step in identifying any changes due to brace wear after a bout of standardized treadmill exercise, and whether changes persist upon the removal of the brace and after 30-minutes of unbraced rest. The first hypothesis predicting changes in force variables, joint angles, joint moments, and EMG variables between unbraced and braced conditions following a treadmill session was nullified suggesting minimal changes due to the brace itself. Second and third hypotheses were subsequently annulled as their prediction involving the changes observed in hypothesis one remaining changed once the brace was removed (hypothesis two) and after 30-minutes of rest (hypothesis three) did not occur. These data suggest that there may not be a risk of injury due to brace-related biomechanical and neuromuscular changes after brace removal as well as after 30-minutes of rest. Though any protective effects were not observed (i.e., changes supporting a decreased risk of injury based upon previously established literature), this work supports the continuation of investigations into the possibility of bracing as a prophylactic adjunct to prevent injury in a high-risk population such as young female athletes.

The methodology used had inherent limitations and therefore may limit the generalizability of the proposed results. First, the task used (single leg landing) is a very standardized task with anticipated movements. It has been shown that females engage in a generalized co-contraction around the knee joint during unanticipated movements (Cowling and Steele 2001). Therefore if a different unanticipated task, like a side-cutting maneuver for example, was implemented into the protocol, there may have been greater differences detected between unbraced and braced conditions. Secondly, a healthy and relatively homogenous cohort was used in this investigation which may have had implications for the results seen. It is likely that expanding the study population to include a previously ACL injured or ACL repaired group would yield different and potentially significant results considering the variables used. However, the healthy population was used to ensure that no confounding variables associated with injury caused changes between sessions and that changes were because of the brace only. Third, with the protocol used, there is no knowledge of what the brace is doing on a mechanical level. Considering the results showing minimal effect of the knee brace on external factors, there is potential that the brace may offer force deflective properties which absorb and deflect excess force causing a difference in loading at the level of the knee joint. Additionally, the brace may have an effect on moment distribution which again would affect the knee joint specifically. An in depth understanding of the mechanical structure of the brace during dynamic loading would have been supremely useful to accurately interpret results in this study beyond the information given using the kinetic, kinematic, and neuromuscular approach. Fourth, no procedure was used to test if there were any proprioceptive effects of the knee brace. This knowledge would have strengthened a few hypotheses in the discussion regarding quicker hamstring onset and increased cocontraction after brace removal. Next, even though care was taken to ensure that the brace fit correctly, there may have been some slipping of the brace during exercise and in turn affecting the position of the shank cluster. This could have had an effect on kinematic calculations and could have been the reasoning for the increased frontal plane joint ROM in the braced versus unbraced condition. Finally, considering the amount of dependent variables presented in this thesis, a correction for multiple comparisons using Bonferroni would provide an alpha level too small to detect statistical significance (i.e., p=0.05/99 = new p-level of 0.0005). In other words, the likelihood of significant main effects due to chance with the amount of comparisons tested is likely equal to the amount of significance actually detected. Therefore interpretation of results should be done so with caution.

An additional limitation during this investigation was the determination of joint moments during the impact phase of landing. Despite the continued use of inverse dynamics to investigate intersegmental loading during movement scenarios, the use of and proper application of signal filtering techniques during impact scenarios is highly debated. It is well established that kinematic data be filtered with a low pass filter at a low cutoff as human motion is often contained under 6 Hz (Winter et al., 1974). However treatment of force plate data, especially in an impact scenario, with the same filtering technique frequently results in severe attenuation of ground reaction forces. It would therefore seem intuitive that each signal is treated appropriately based on their subsequent frequency contents, however this is not necessarily the case. The eventual multiplication of two signals treated with differing frequency contents tends give what is considered as an unphysiological artifact that occurs upon force plate contact. One of the first papers to identify this issue was van den Bogert and de Koning (2006) who demonstrated with an optimization analysis that large errors occurred when kinematics and kinetics were filtered with different frequencies due to a combination of impact peaks in the horizontal ground reaction force and the inability to calculate the frequency components of segmental accelerations. They concluded that data be treated dependent on the variable of interest, namely if ground reaction forces are the variable of interest use a cutoff that is appropriate for the frequency content and if moments are required us a low and consistent cutoff for both signals in order to attenuate the artifact (van den Bogart & de Koning 1996). Since this publication, there have been several investigations stating similar results (Bisseling & Hof 2006; Kristianlund et al, 2011). Additionally, Kristianlund et al (2011) found that filtering procedure affected the ranking of at-risk athletes based on knee abduction moment in a large sample of handball players (N=123). Therefore, filtering both the kinematic data and force plate data at the same low pass cutoff frequency is a common method to try and minimize artifact at impact.

Despite the evidence presented above, some authors continue to use different filtering procedures for each of the kinematic and kinetic data sets (Hewett et al., 2005; Laughlin et al., 2011; Sigward & Powers 2007). The implications for uncertainly in data treatment for inverse dynamics in an impact scenario is concerning for ACL injury research as interpretation of data related to movement patterns and injury risk may be adversely affected by errors in data treatment techniques. It is likely that the reason current literature tends to steer toward posture and joint kinematics in identifying high risk individuals (as in Hewett et al., 2005) is a result of this conflict. Considering the evidence provided above, the interpretation of brace wear joint moments with this dataset must be done so with extreme caution. As well, troubles in the literature regarding joint moments during impact may in turn affect the interpretation of interaction effects observed for joint moments. There was an interaction effect for sagittal knee moment at peak force for the third hypothesis (after 30-minutes of rest) with no significance of session at either time point (4 or 5). This is potentially due to the interaction effect seen with the sagittal plane knee angle at peak force and unchanging peak Fz variable as joint moment is a function of changing moment arms with changing joint angle and force. Further, looking at the trends of the mean differences, one can see an opposite trend from the sagittal knee angle at peak force as differences between braced and unbraced conditions are larger for time 4 and start to come back to parity at time 5. This may suggest that the reason the participants during the braced session landed with a higher degree of knee flexion at peak force was to bring sagittal knee moments closer to a pre-activity level (as further suggested by the nonsignificant post-hoc test done in the overall ANOVA ($p=0.220$)).

Chapter 6: Thesis Synthesis and Conclusions

The purpose of this thesis work was to characterize biomechanical and neuromuscular changes associated with prophylactic knee brace wear before and after standardized exercise, after brace removal, and 30-minutes after the cessation of exercise. This data would provide baseline evidence to answer whether there is an acute level of brace acclimatization with exercise and whether any documented changes last after the brace is removed. Further, this work would set a foundation for potential future work in longitudinal brace wear over repeated bouts of brace wear during exercise over time. A total of 99 ANOVAs were run for force, joint angle, joint moment, and EMG dependent variables over three hypotheses. Only four main effects of brace and seven interaction effects of time and brace were observed from these dependent variables. Due to the large number of tests run, a statistical adjustment of alpha for multiple comparisons would have created a new alpha level that was too small to detect significant changes. Therefore, the main effects and interaction effects observed could have been due to chance and were difficult to interpret with accuracy. Additionally, equivalency of means results demonstrated that variables tended to change as a function of time (exercise), not a function of brace. A sub-analysis identified two participants that may be at a higher risk of ACL injury due to their kinematics and neuromuscular variables during unbraced landing. It was found that the brace changed frontal knee positioning to a more varus alignment during landing which is thought to be more protective. As well, for one of the high risk participants, the brace decreased pVGRF, increased sagittal plane joint ROM, resulted in quicker gluteus medius onset, and decreased muscle cocontraction at the knee. The results from the sub-analysis suggest a neuromuscular effect (though unable to assess statistically) by changing joint positioning and in turn affecting muscular activity. It was concluded in this work that the prophylactic knee brace had minimal biomechanical and neuromuscular effects on healthy university-aged active female participants following exercise.

The results of this work demonstrated that the CTi OTS prophylactic brace had minimal effect on loading, joint angle, joint moment, muscle onset, and co-contraction variables during single leg landings before and after a treadmill intervention. Although a lack of significance was not expected, it may have been due to a few limitations. First, though the single leg landing procedure is a documented screening test in the literature able to detect differences between braced and unbraced conditions (Rishiraj et al., 2012) a procedure involving unanticipated movements may have demonstrated effects more in-line with my hypotheses. Secondly, the population tested was healthy and relatively homogenous. There is potential that using a previously injured populations or a population pre-screened for injury risk could have had an effect on the results. Finally, a major limitation in this work is the use of joint moments in any data interpretation as the filtering of two different signals causes unphysiological artifacts in the data during impact. As intersegmental joint loading is an important consideration in injury risk with respect to ACL injury and bracing, it seems wise that further rigourous investigation into filtering techniques with impact data is required to parse out signal artifact from physiological motion and will allow for better interpretation of data. This would strengthen current screening protocols for athletes to determine those at a high risk of future knee injury.

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Appendices

Appendix A: Major and Minor Ligaments of the Knee Complex

Table A-1: Major and minor structures of the knee joint. From Moore and Daley 2006.

Major Ligaments	Location	Function
Medial Collateral Ligament (MCL)	The ligament attaches on the medial femoral condyle and inserts on the medial condyle of the tibia	Resists valgus motion of the tibia upon the femur
Lateral Collateral Ligament (LCL)	The ligament attaches on the lateral femoral condyle and inserts on the lateral condyle of the tibia	Resists varus motion of the tibia upon the femur
Anterior Cruciate Ligament (ACL)	The ligament attaches on the posterior aspect of the distal femur and inserts on the anterior aspect of the tibial plateau. The posterolateral and anteromedial bands are named as such depending on what area of the tibial plateau they attach.	Primary function is to resist anterior tibial translation of the tibia upon the femur. The secondary function is to prevent internal rotation of the tibia.
Posterior Cruciate Ligament (PCL)	The ligament attaches on the antero-lateral aspect of medial femoral condyle in the area of intercondylar notch and inserts on the posterior aspect of the tibial plateau.	Resists posterior tibial translation of the tibia upon the femur.
Minor Ligaments	**Location**	**Function**
Oblique Popliteal Ligament	The ligament arises posterior to the medial tibial condyle and passes superolaterally toward the lateral femoral condyle, blending with the posterior part of the joint capsule.	An expansion of the semimembranosus tendon that reinforces the joint capsule posteriorly.
Arcuate Popliteal Ligament	The ligament arises from the posterior aspect of the fibular head, passes superomedially over the tendon of the popliteus, and spreads over the posterior surface of the knee.	Strengthens the capsule posterolaterally.
Transverse Ligament of the Knee	The ligament joins the anterior edges of the menisci.	Tethers the mensci during knee movement.
Coronary Ligament	Connects the inferior edges of the menisci to the tibia.	Connect the menisci to the joint capsule

Appendix B: Electromyography Normalization Procedures

Table B-1: Normalization procedures for EMG of the lower limb musculature

Muscle	Description of Placement	Description of Normalization Procedure
Biceps femoris	Along the line from the ischial tuberosity to the lateral aspect of the popliteal fossa. Staring from the ischium, place electrode approximately 1/3 of the distance along the anatomical line. Electrode should be oriented on a slight slant running in the direction of the muscle fibres.	Have the participant lay prone on the massage table in the lab space. Bend the knee to 90 degrees and have the researcher support the low back and the ankle of the bent leg. Have the participant ramp into a maximum voluntary contraction by asking the participant to try to bring their shank to the back of their thigh.
Rectus femoris	Starting in the middle of the patella, draw an anatomical line to the ASIS. Electrode placement should be approximately halfway along the line. Be wary of the position of the brace to ensure that it will not be covered.	Have the participant sitting up with their legs hanging over the edge of the table. Have the secondary researcher support the upper back of the participant. The other researcher will bring the leg to 90 degrees then brace the front of the distal part of the shank. The participant will be asked to ramp into a maximum voluntary contraction by extending the knee against resistance.
Gastrocnemius	Placement is slightly medial, and just below the bottom band of the brace. This placement is not ideal; however need to accompany the brace.	A fabricated shrug board will be used to elicit the maximum voluntary contraction of the gastrocnemius muscles. The participant’s feet will be strapped into the board with adjustable straps, followed by attached shoulder straps adjusted snugly around the shoulders. The participant will be asked to push up onto their tip toes.
Gluteus medius	Get the participant, while standing, to ‘spread the floor’. Once the anticipated spot is covered, get the participant to extend the leg -45 degrees to ensure position. Orient the electrode along an anatomical line from the greater trochanter to the iliac crest.	Have the participant lay on their side on the massage table. Take the top leg and bend the hip so that the top foot rests on the knee of the bottom straight leg. The researcher will stand on the posterior side of the participant and place a hand on the knee of the top leg. The participant will then be asked to ramp into a maximum voluntary contraction by externally rotate their hip into the resistance of the researchers hand.

Appendix C: Landmarking Procedures

Table C-1: Landmarking for the thigh rigid body cluster

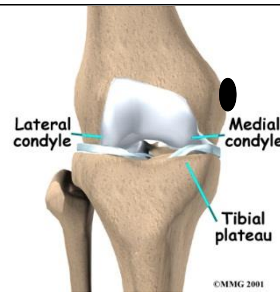
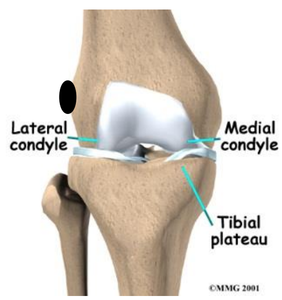
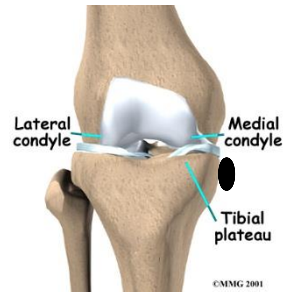
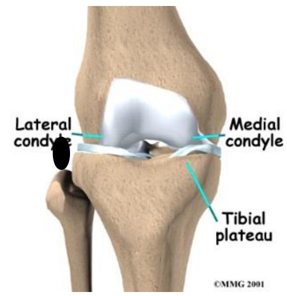
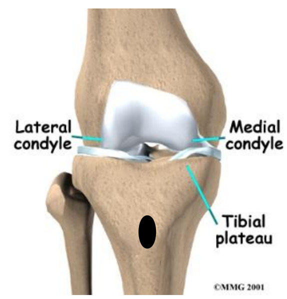
Thigh Cluster			
Landmark	Marker	Palpation Technique	Picture
Greater trochanter	RGT/ LGT	See Lower Back Cluster	See Lower Back Cluster
Medial Femoral Condyle	RFMC/ LFMC	• Feel for the tibial tuberosity on the anterior surface of the proximal tibia • Guide your fingers medially along the joint line • Ask the participant to bend the knee • Move the fingers proximally above the joint line to the medial bony landmark of the medial condyle	
Lateral Femoral Condyle	RFLC/ LFLC	• Feel for the tibial tuberosity on the anterior surface of the proximal tibia • Guide your fingers laterally along the joint line • Ask the participant to bend the knee forward • Move the fingers proximally above the joint line to the lateral bony landmark of the lateral condyle	

Table C-2: Landmarking for the shank rigid body cluster

Shank Cluster			
Landmark	Marker	Palpation Technique	Picture
Medial Tibial Condyle	RMTC/ LMTC	• Feel for the tibial tuberosity on the anterior surface of the proximal tibia • Guide your fingers medially along the joint line • Ask the participant to bend the knee • Ensure to place the probe below the joint line along the condyle	
Lateral Femoral Condyle	RFLC/ LFLC	• Feel for the tibial tuberosity on the anterior surface of the proximal tibia • Guide your fingers laterally along the joint line • Ask the participant to bend the knee forward • Feel for the head of the fibula • Place the probe above the head of the fibula and below the joint line	
Tibial Tuberosity	RTIB/ LTIB	• Find the sharp protuberance located on the anterior surface of the proximal tibia, just below the patella	
Medial Malleolus	RMM/ LMM	• Locate the bony protuberance on the medial side of the distal tibia • ‘Ankle bone’	

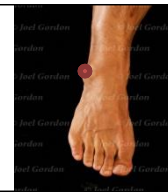
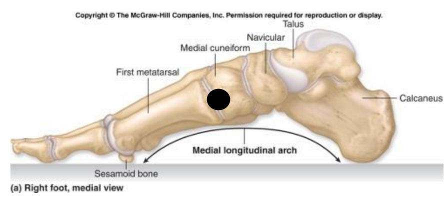
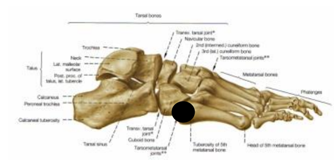
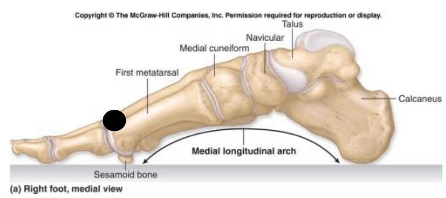
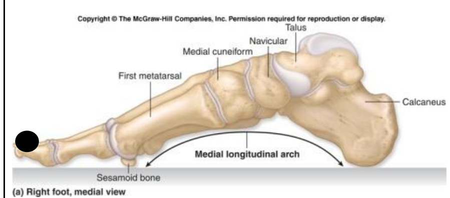
Lateral Malleolus	RLM/ LLM	• Locate the bony protuberance on the lateral side of the distal tibia • ‘Ankle bone’	

Table C-3: Landmarking for the food rigid body cluster

Foot Cluster			
Landmark	Marker	Palpation Technique	Picture
Medial Malleolus	RMM/ LMM	See Shank Cluster	
Lateral Malleolus	RLM/ LLM	See Shank Cluster	
Superior Midfoot	RSMF/ LSMF	• Start with your fingers on the medial malleolus • Move longitudinally down the foot to the first protuberance (talus) • Continue past the next protuberance (navicular) and find the third protuberance (cuniform)	
Lateral Midfoot	RLMF/ LLMF	• Locate the proximal protuberance of the fifth metatarsal where it junctions with the cuboid bone	
Heel	RHEEL/ LHEEL	• Place the probe at the most posterior aspect of the calcaneous	
First Metatarsal	RMT1/ LMT1	• Locate the head of the first metatarsal • Base of the 'big toe'	
First Toe	RTOE/ LTOE	• Place the marker (for the right foot) and the probe (for the left foot) on top of the big toe	

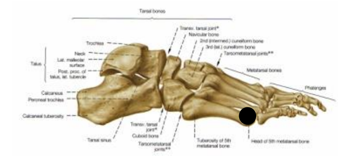
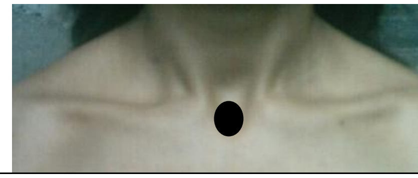
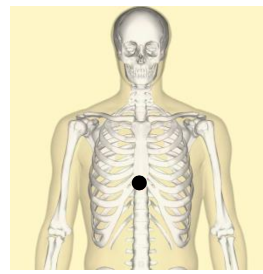
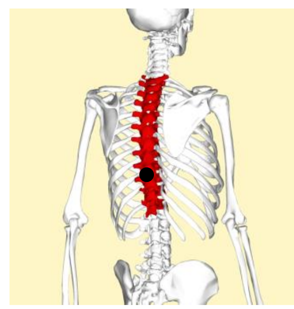
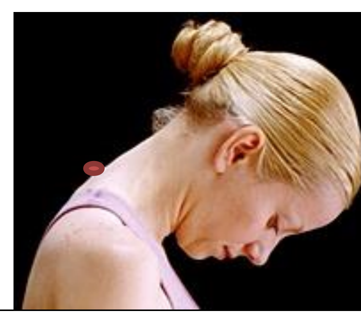
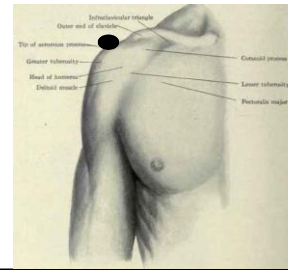
Fifth Metatarsal	RMT5/ LMT5	• Locate the head of the fifth metatarsal • Base of the 'baby toe'	

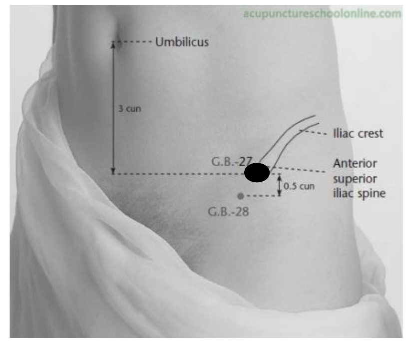
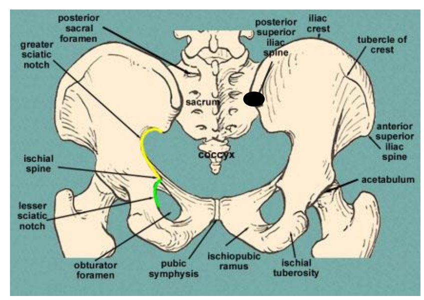
Table C-4: Landmarking for the upper back rigid body cluster

Upper Back Cluster

Landmark	Marker	Techniques for Palpation	Picture
Suprasternal Notch	SS	• Locate the space between the medial ends of the clavicles • Place the end of the probe in this notch	
Xiphoid Process (note: this is a very sensitive area which should NOT be harshly palpated)	XP	• Locate the bottom of the ribcage on either side (12th rib) • Follow the ribcage medially until your fingers meet in the middle	
T10	T10	• Find the inferior angle (IA) of the scapula on one side • The spinous process immediately medial to the IA is approximately T8 • Palpate 2 spinous processes inferiorly	
C7	C7	• Ask the participant to look toward the floor • Examine the spinous process that projects from the posterior aspect of the neck (C7) • Place the probe on this spinous process	

Acromion (left and right)	RAC/LAC	• Palpate the clavicle laterally until you reach the acromioclavicular joint • The acromion is just lateral to the joint capsule and feels like a ‘shelf’	

<u>Table C-5:</u> Landmarking for the low back rigid body cluster

Low Back Cluster			
Landmark	Marker	Techniques for Palpation	Picture
Anterior superior iliac spine	RASIS/ LASIS	• From the anterior side of the participant, use the hands to locate the tops of the iliac crests • With the medial sides of the index fingers on top of the crests, allow the thumbs to run along the crests and down to the protuberance of the ASIS	
Posterior superior iliac spine (note: will be helpful to palpate this before the application of the low back cluster)	RPSIS/ LPSIS	• From the posterior side of the participant, use the medial sides of the index fingers to locate the top of the iliac crest • Find where the sacrum meets the pelvis (sacroilio joint; 'Dimples of Venus') • Follow the sacrum posteriorly to the protuberances of the PSIS	
Greater trochanter (note: make sure you palpate the bone and not the muscle)	RGT/ LGT	• Ask the participant to pretend like they're 'squashing a bug' with their toe • Feel for the head of the femur (i.e., greater tubercle of the trochanter) turn under the skin	

Appendix D: Lab Set-Up

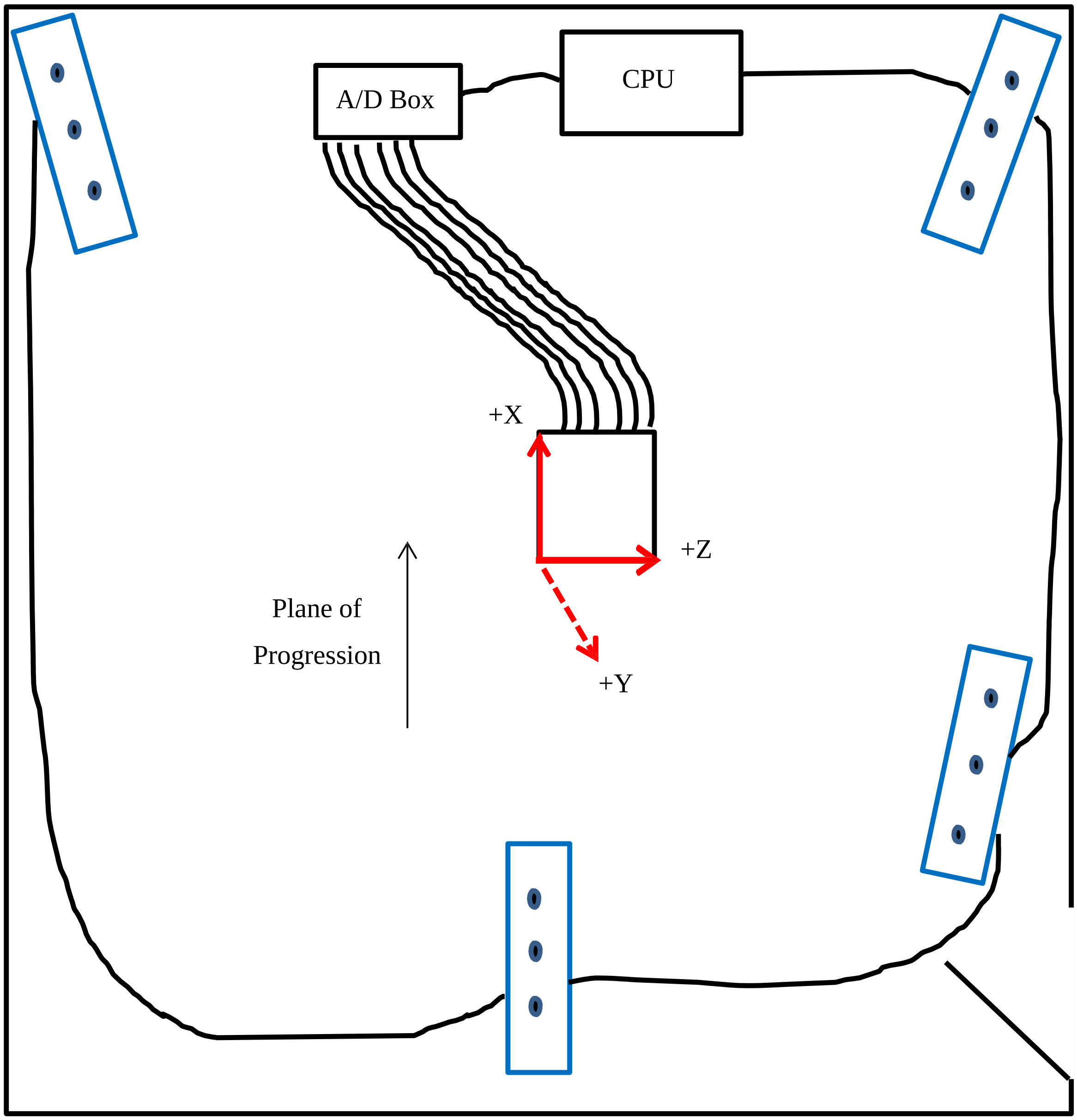


Figure D-1: Schematic diagram of the IBAL lab space.

Appendix E: Medical Questionnaire

Medical History

Please indicate with a checkmark if you currently have or previously had any of the following conditions:

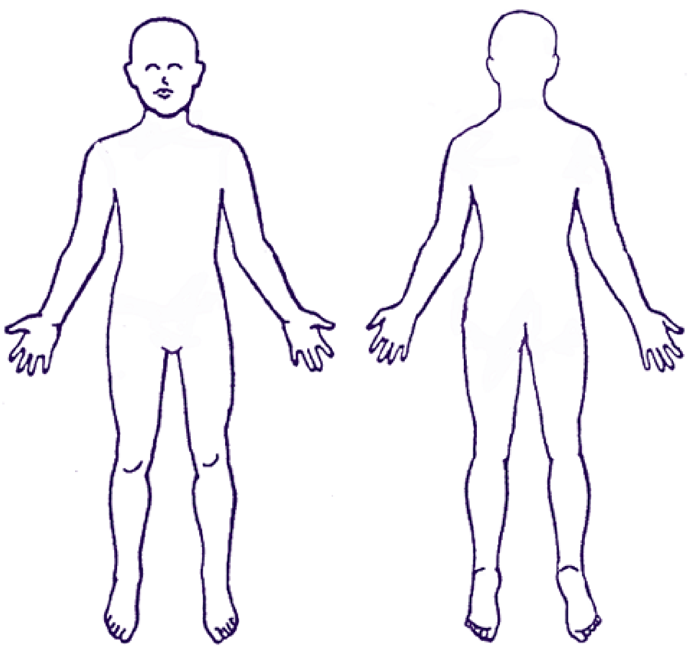
- Diabetes
- Hepatitis
- Liver disease
- Thyroid problems
- Respiratory problems
- Allergies
- Hay fever
- Sinus problems
- Arthritis
- Kidney trouble
- Tuberculosis
- Epilepsy
- Immune problems
- Glaucoma
- Cardiac problems
- Low blood pressure

Do you have any other contraindications to exercise? If yes, please explain:______________________

____

____

Please mark on the body with circle previous musculoskeletal injuries, indicate the approximate date of injury, and any lasting effects that may affect your ability to participate in this study:



____

____

If you have suffered knee injuries in the past, please fill out this section. If not, move on to the "Activity History" portion.

Indicate with a checkmark if you currently experience any of the following symptoms in your ***affected*** knee:

- Pain
- Instability
- Other:____________
- Swelling
- Locking
- Crepitus ('cracking')

Indicate any activities that exacerbate any of the above symptoms:

- Walking
- Biking
- Other:____________
- Running
- Jumping
- Stairs
- Agility

Activity History

How many times per week do you engage in moderate-to-vigorous physical activity?:

- 1-2
- 3-5
- 6-7+

Please indicate the types of activities you often engage in:

- Cycling
- Yoga/Pilates
- Dance
- Step/Boot Camp
- Running
- Interval Training
- Weight Lifting
- Agility Training
- Racquet Sports
- Skating
- Basketball
- Volleyball
- Ice Hockey
- Soccer
- Martial Arts
- Ultimate Frisbee
- Football
- Other:____

Appendix F: Knee Brace Subjective Questionnaire

Knee Brace Subjective Comfort Questionnaire

Subjective scales are used in research to give the investigators an idea of what the participant is feeling during the collection of physical data or through an intervention. Using the words below, please circle a numerical category that describes ***THE COMFORT OF THE BRACE ONLY*** as it pertained to this exercise session. Please be accurate in your descriptions. An example of how to fill out the questionnaire is provided below. If there are any questions, please ask the investigators.

Example:

On the scale below, please indicate the level of fatigue reached during this exercise session:

None at all 0 1 2 3 4 5 Unbearable fatigue

Question 1: Please indicate the level of slippage experienced during the duration of the exercise session:

None at all 0 1 2 3 4 5 Constantly slipped

Question 2: Please indicate the overall comfort of this brace:

Not comfortable 0 1 2 3 4 5 Very comfortable

Question 3: If you were told that this brace may prevent knee injuries during sporting activities, would you choose to wear the brace?

Not at all 0 1 2 3 4 5 Absolutely

Appendix G: Borg 6-20 RPE Scale

RPE Scale

rating of perceived exertion

rating	description
6	NO EXERTION AT ALL
7	EXTREMELY LIGHT
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

for more information see http://www.topendsports.com/testing/rpe-scale.htm

Appendix H: Rate of Perceived Discomfort (Knee Brace)

Rate of Perceived Discomfort

Please indicate on the scale by circling the number correlated to the amount of current comfort/discomfort you are feeling with the application of the knee brace.

10 *maximum comfort*
9
8
7
6
5 *strong comfort*
4
3 *moderate comfort*
2 *weak comfort*
1
0.5 *noticeable discomfort*
0
–0.5 *just noticeable discomfort*
–1
–2 *weak discomfort*
–3 *moderate discomfort*
–4
–5 *strong discomfort*
–6
–7
–8
–9
–10 *maximum discomfort*

(From Hernandez et al., 2002)

Appendix I: Variable Means and Variances

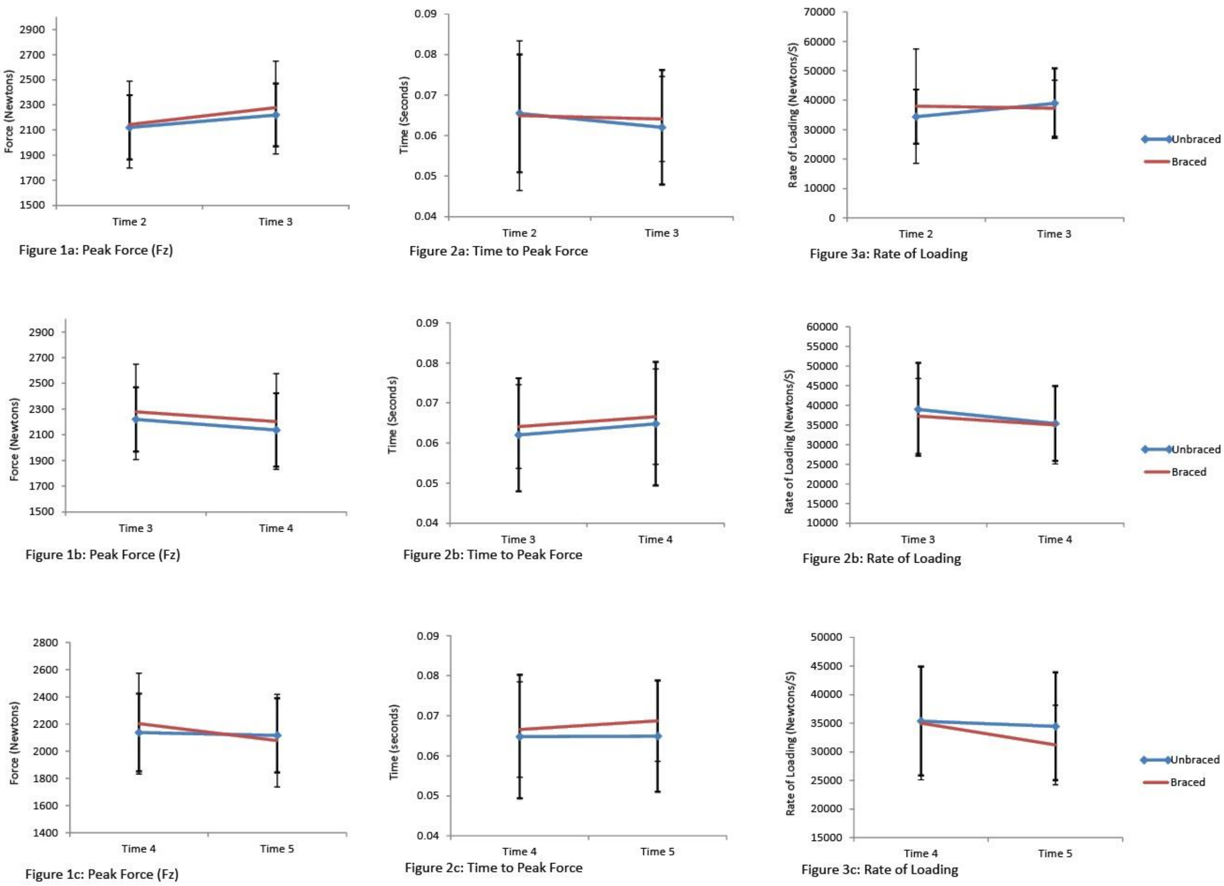


Figure I-1: Graphed means and variances for peak Fz force, time to peak Fz, and rate of loading variables for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

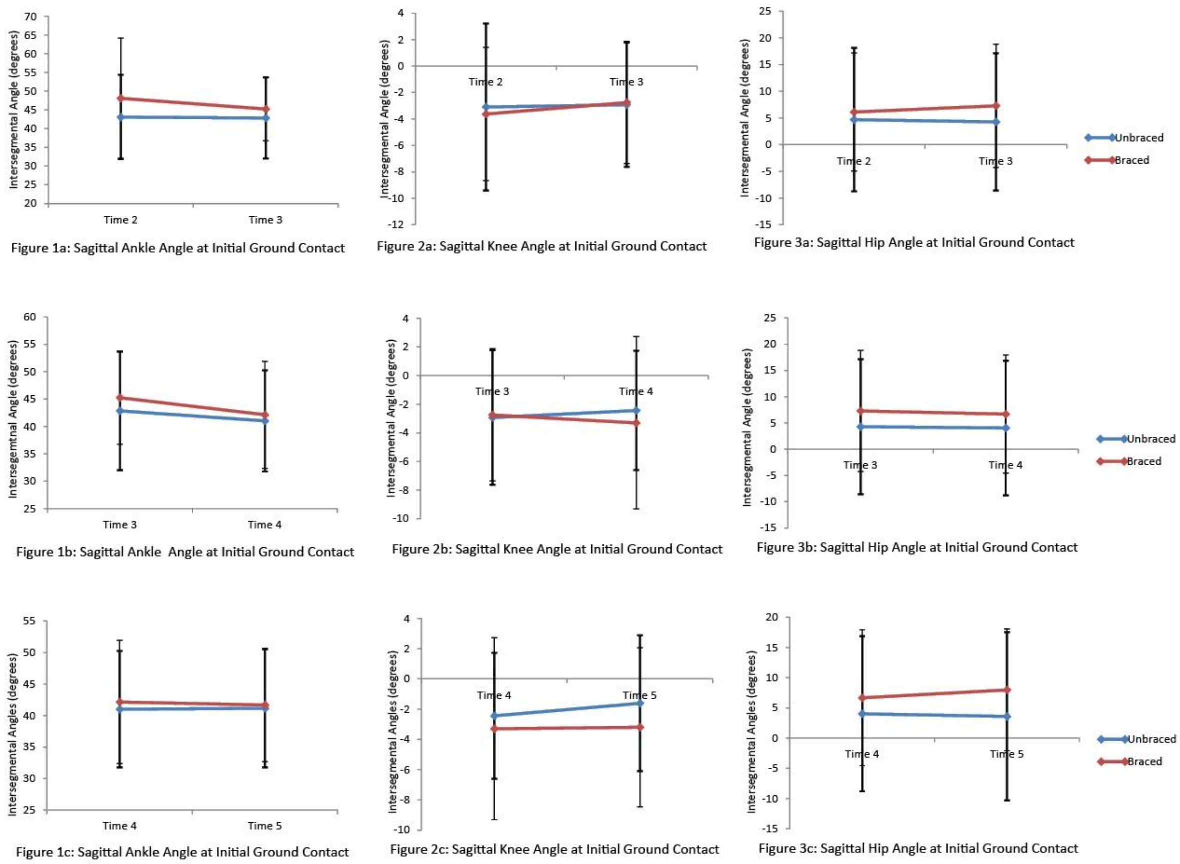


Figure I-2: Graphed means and variances for ankle, knee, and hip joint angles at initial force plate contact variables for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

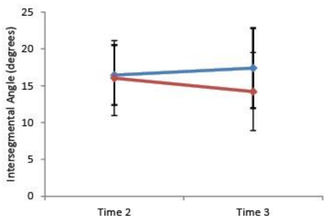


Figure 1a: Frontal Ankle Angle at Initial Ground Contact

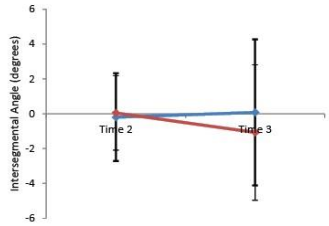


Figure 2a: Frontal Knee Angle at Initial Ground Contact

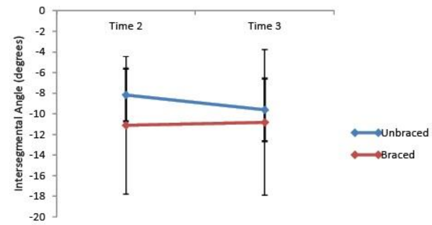


Figure 3a: Frontal Hip Angle at Initial Ground Contact

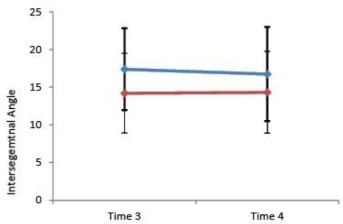


Figure 1b: Frontal Ankle Angle at Initial Ground Contact

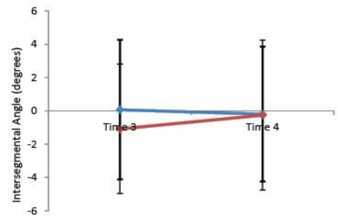


Figure 2b: Frontal Knee Angle at Initial Ground Contact

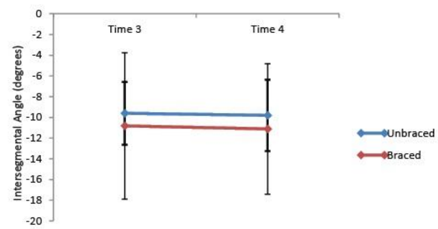


Figure 3b: Frontal Hip Angle at Initial Ground Contact

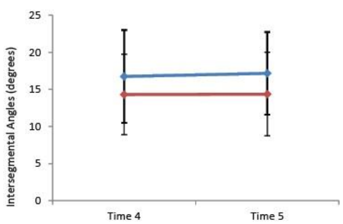


Figure 1c: Frontal Ankle Angle at Initial Ground Contact

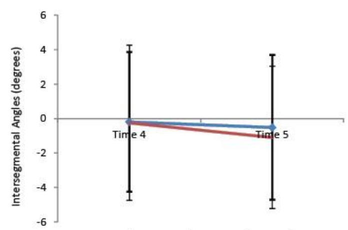


Figure 2c: FrontalKnee Angle at Initial Ground Contact

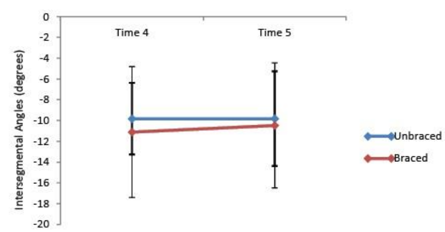


Figure 3c: Frontal Hip Angle at Initial Ground Contact

Figure I-3: Graphed means and variances for ankle, knee, and hip frontal plane joint angles at initial force plate contact variables for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

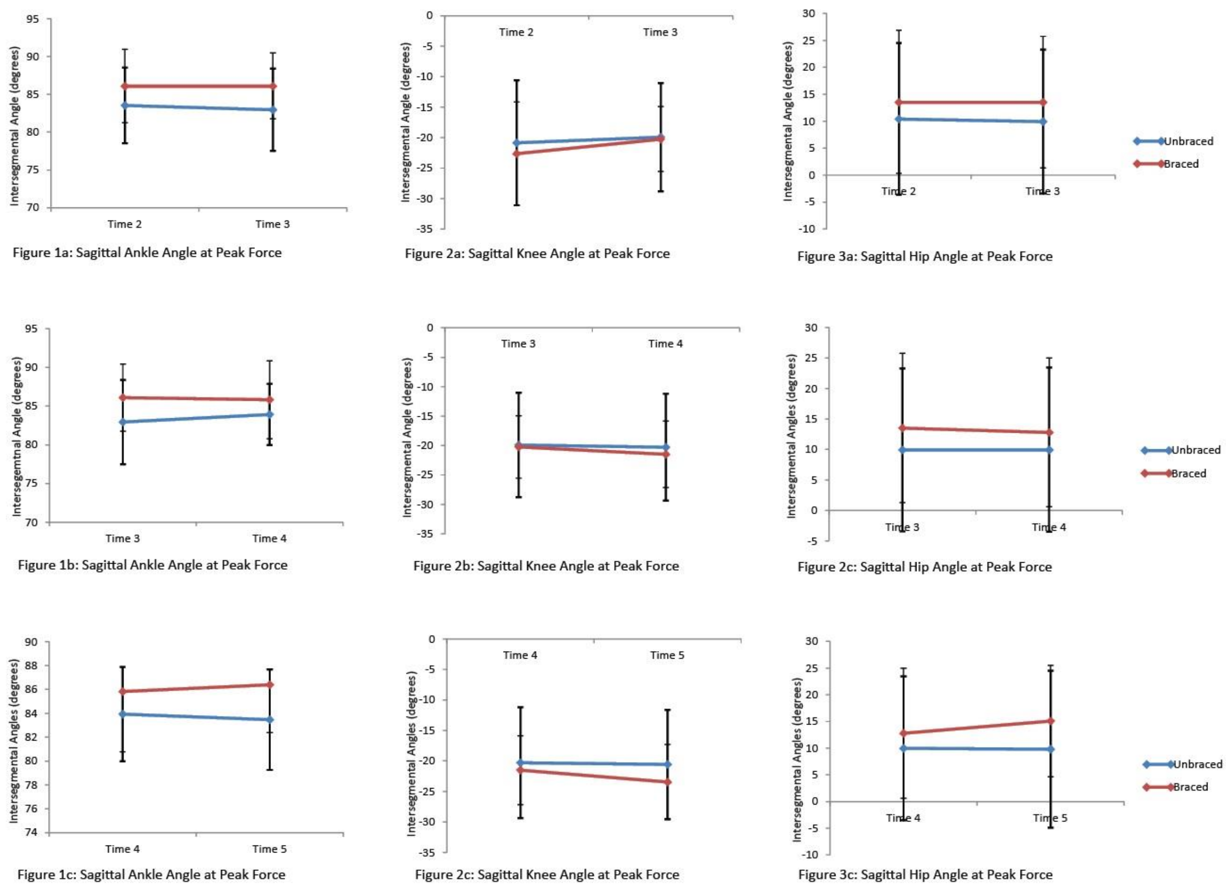


Figure I-4: Graphed means and variances for ankle, knee, and hip plane joint angles at peak Fz force for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

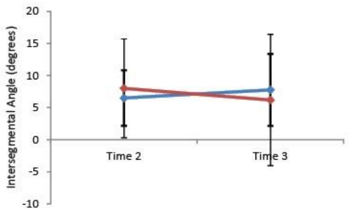


Figure 1a: Frontal Ankle Angle at Peak Force

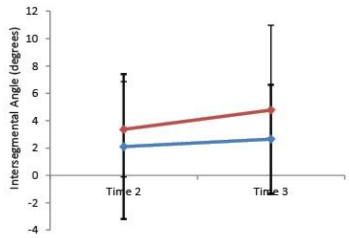


Figure 2a: Frontal Knee Angle at Peak Force

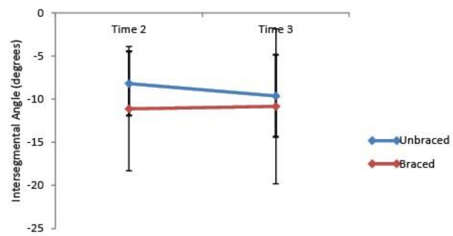


Figure 3a: Frontal Hip Angle at Peak Force

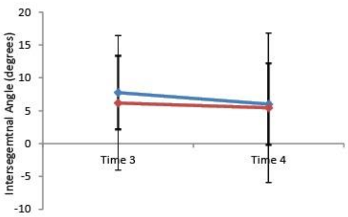


Figure 1b: Frontal Ankle Angle at Peak Force

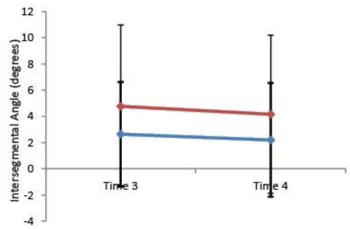


Figure 2b: Frontal Knee Angle at Peak Force

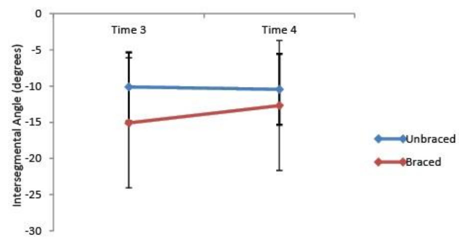


Figure 3b: Frontal Hip Angle at Peak Force

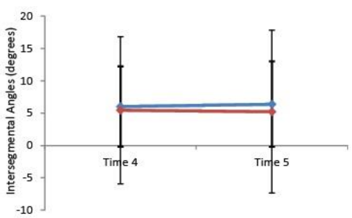


Figure 1c: Frontal Ankle Angle at Peak Force

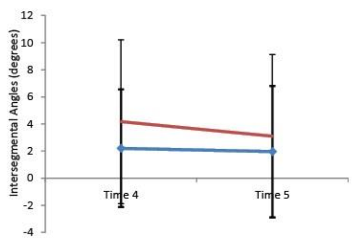


Figure 2c: Frontal Knee Angle at Peak Force

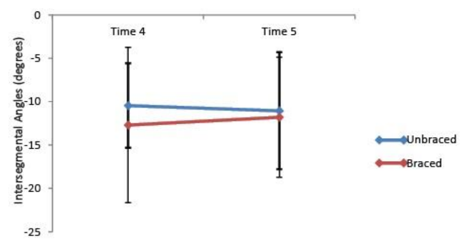


Figure 3c: Frontal Hip Angle at Peak Force

Figure I-5: Graphed means and variances for ankle, knee, and hip frontal plane joint angles at peak Fz force for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

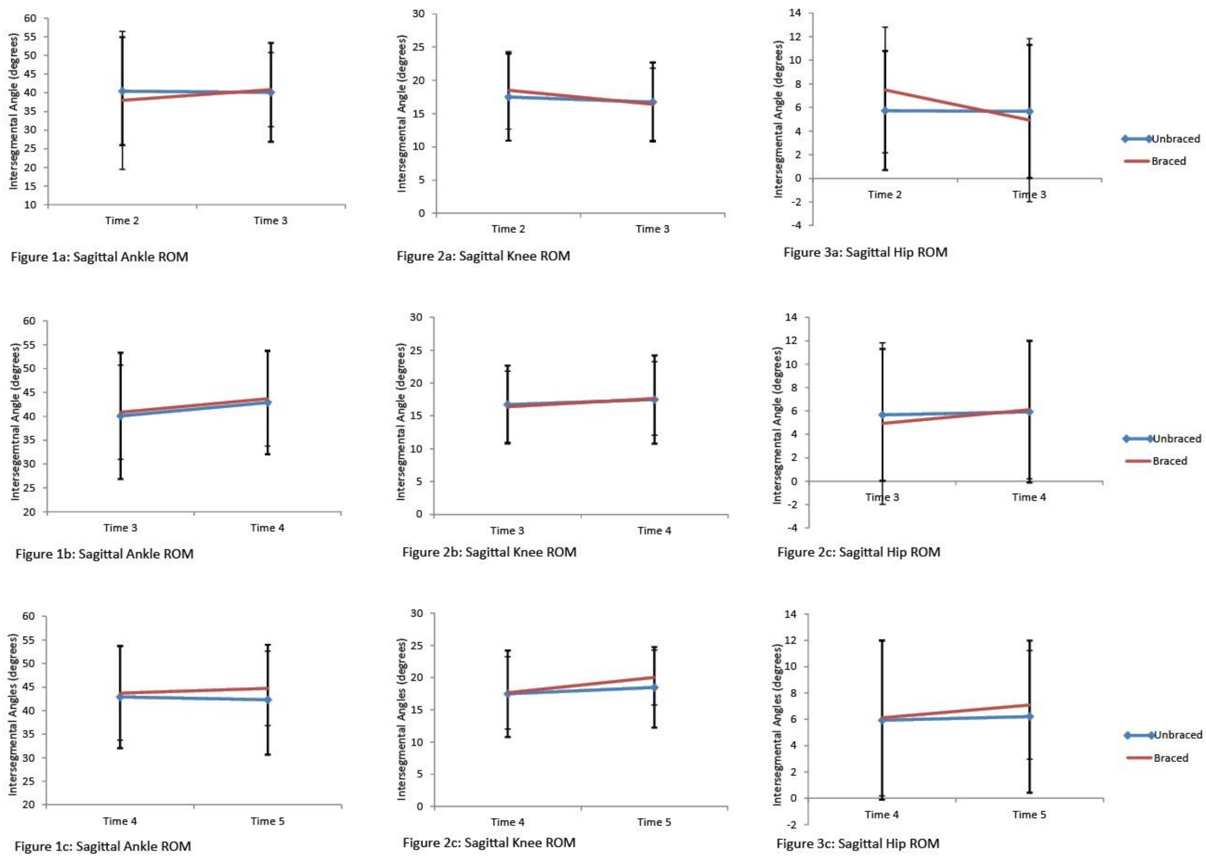


Figure I-6: Graphed means and variances for ankle, knee, and hip sagittal plane joint angle range of motion (ROM) between force plate contact and peak Fz force for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

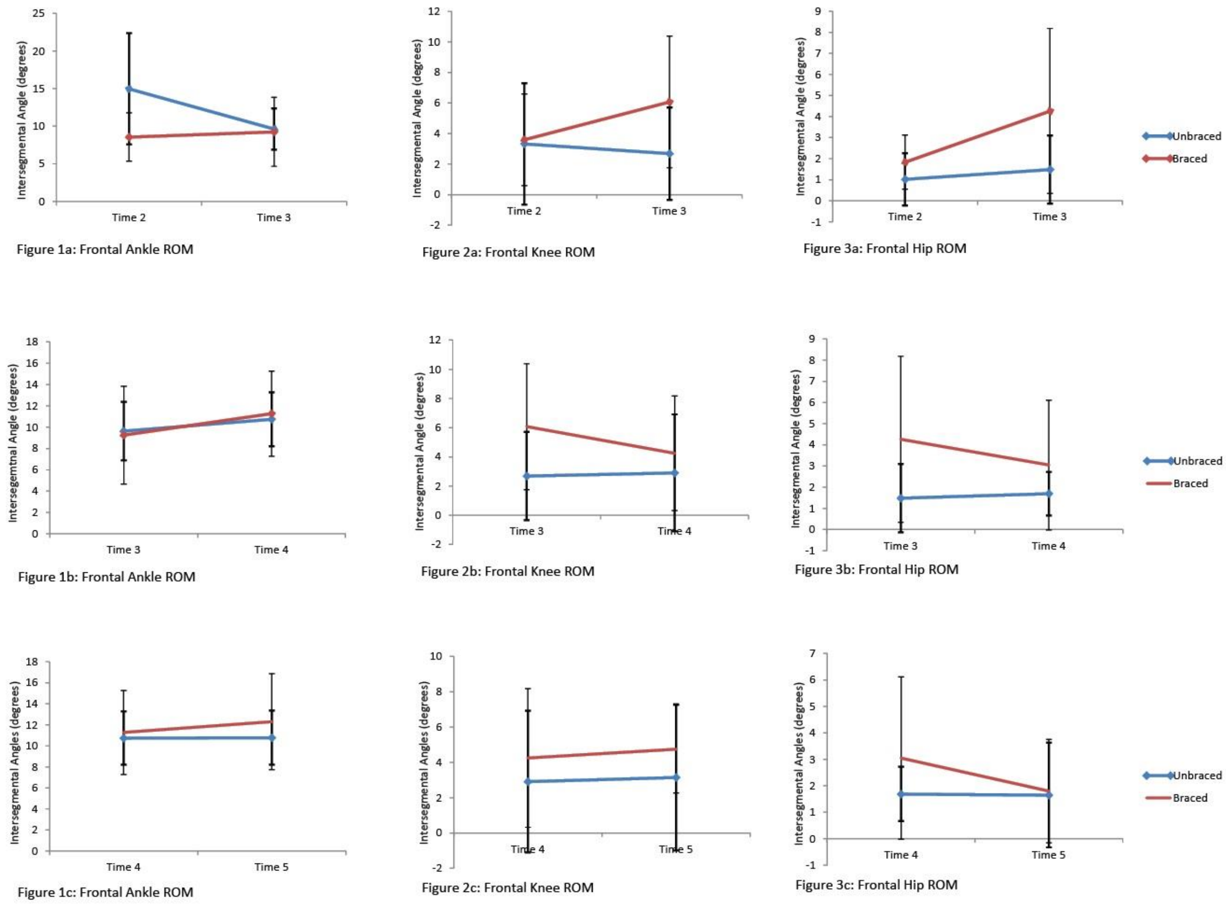


Figure I-7: Graphed means and variances for ankle, knee, and hip frontal plane joint angle range of motion (ROM) between force plate contact and peak Fz force for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

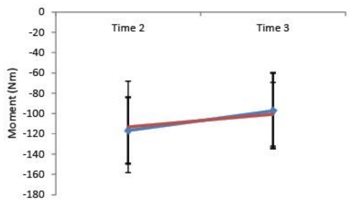


Figure 1a: Sagittal Ankle Moment at Peak Force

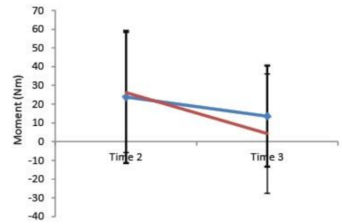


Figure 2a: Sagittal Knee Moment at Peak Force

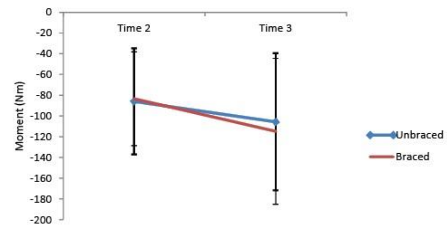


Figure 3a: Sagittal Hip Moment at Peak Force

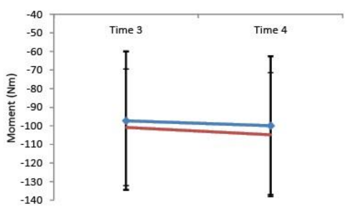


Figure 1b: Sagittal Ankle Moment at Peak Force

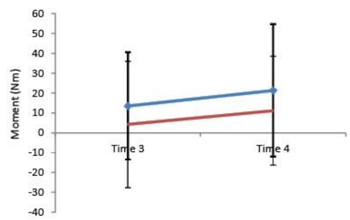


Figure 2b: Sagittal Knee Moment at Peak Force

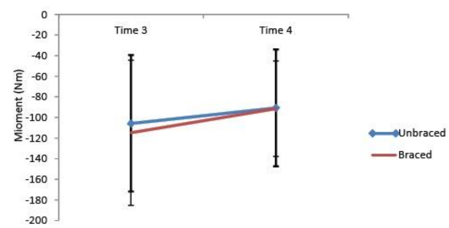


Figure 2c: Sagittal Hip Moment at Peak Force

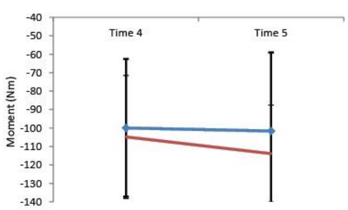


Figure 1c: Sagittal Ankle Moment at Peak Force

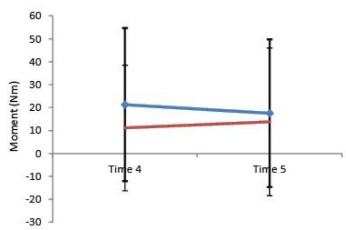


Figure 2c: Sagittal Knee Moment at Peak Force

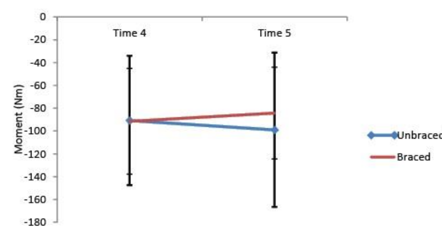


Figure 3c: Sagittal Hip Moment at Peak Force

Figure I-8: Graphed means and variances for ankle, knee, and hip sagittal plane moment at peak Fz force for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

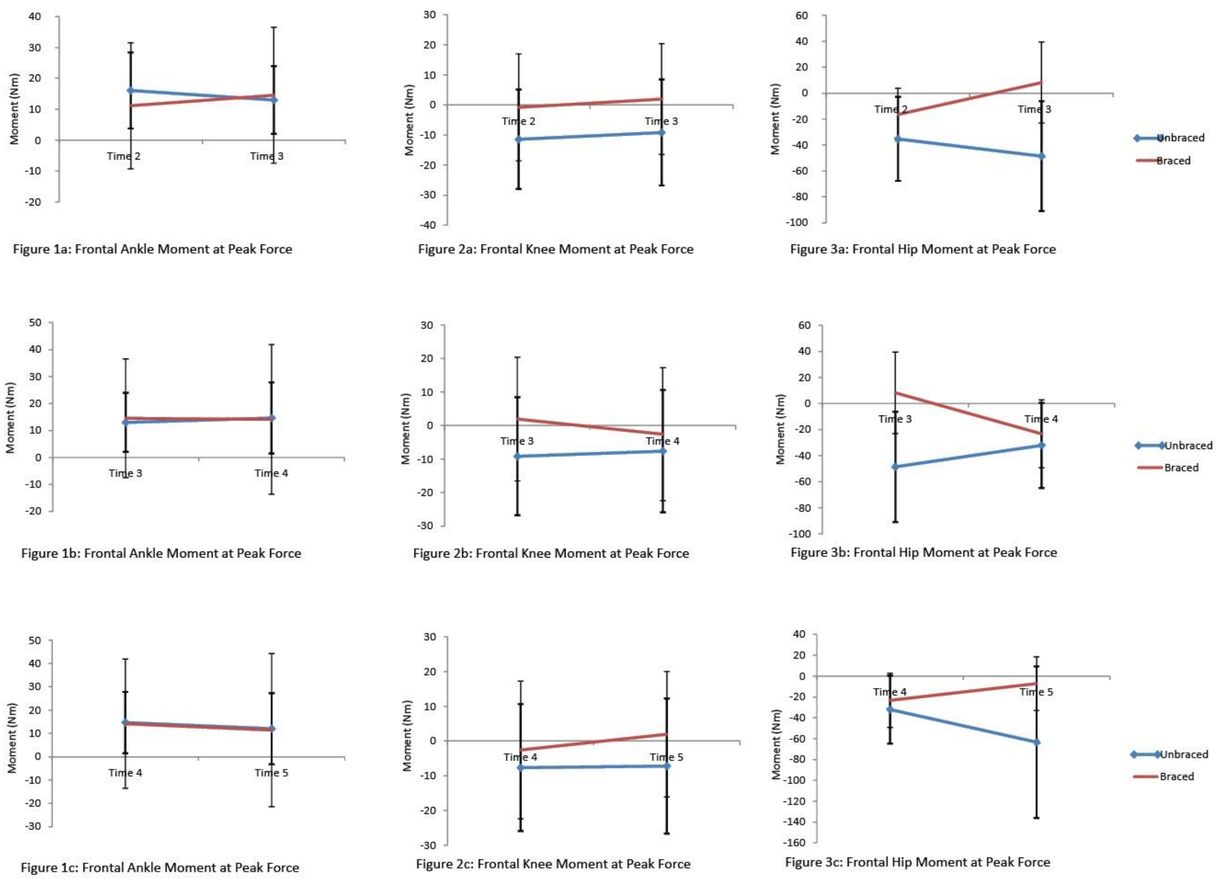


Figure I-9: Graphed means and variances for ankle, knee, and hip frontal plane moment at peak Fz force for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

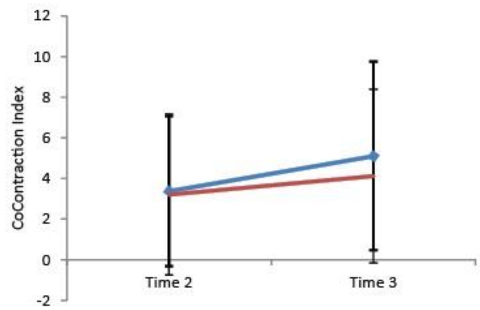


Figure 1a: Preparatory Phase CCI RF/BF

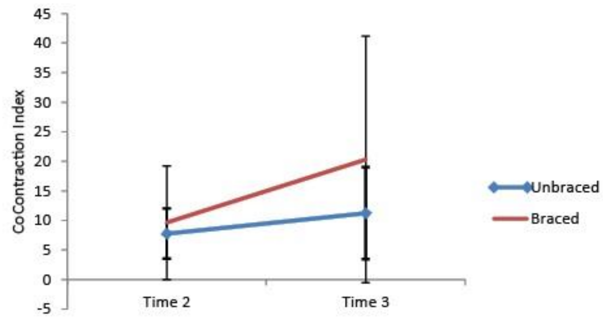


Figure 2a: Preparatory Phase CCI RF/G

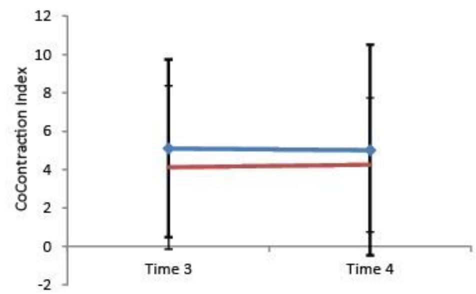


Figure 1b: Preparatory Phase CCI RF/BF

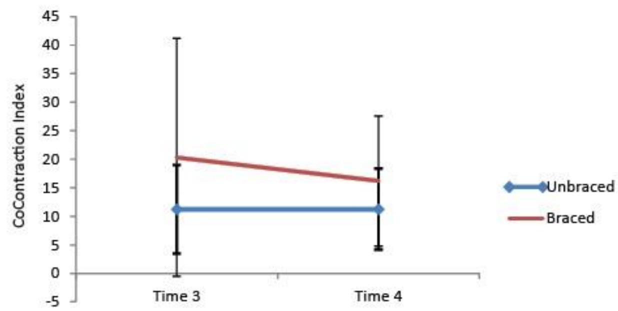


Figure 2b: Preparatory Phase CCI RF/G

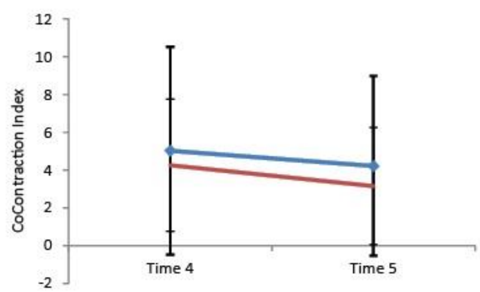


Figure 1c: Preparatory Phase CCI RF/BF

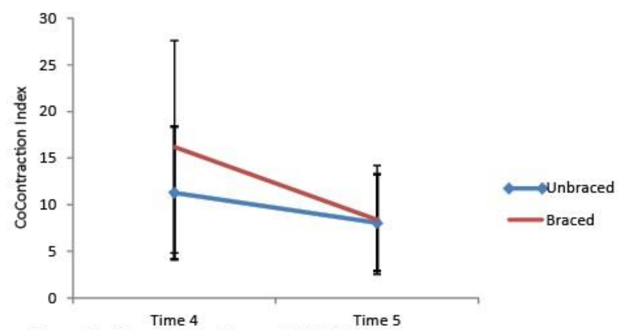


Figure 2c: Preparatory Phase CCI RF/G

Figure I-10: Graphed means and variances for preparatory phase co-contraction index variables for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

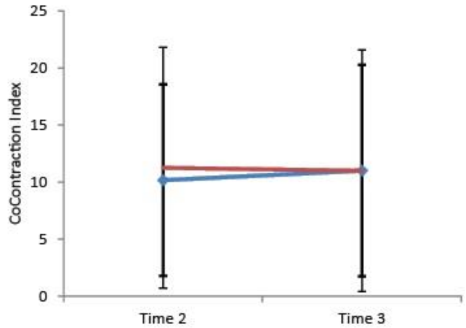


Figure 1a: Landing Phase CCI RF/BF

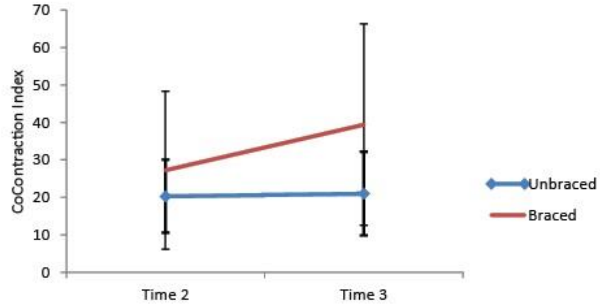


Figure 2a: Landing Phase CCI RF/G

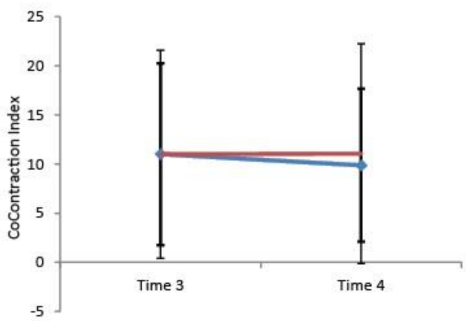


Figure 1b: Landing Phase CCI RF/BF

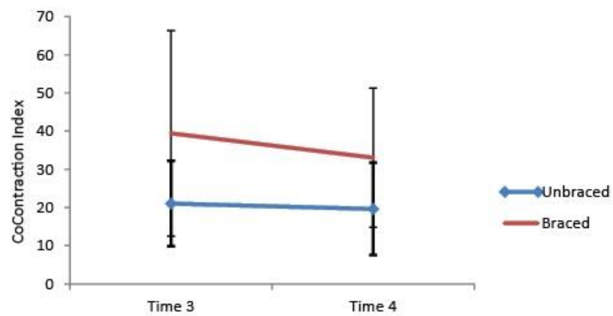


Figure 2b: Landing Phase CCI RF/G

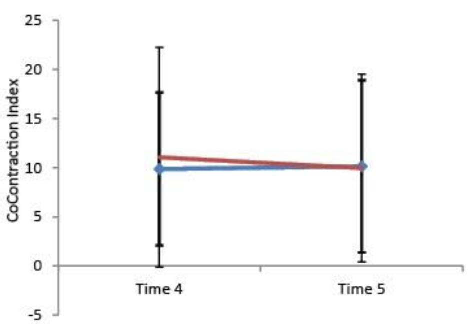


Figure 1c: Landing Phase CCI RF/BF

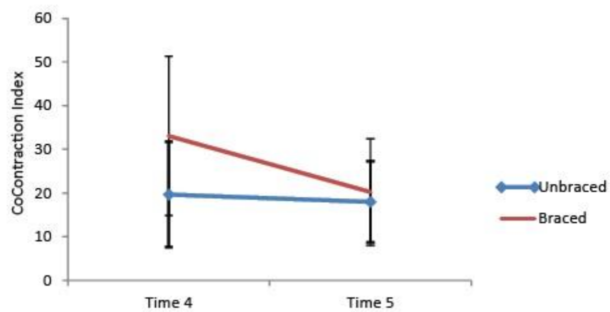


Figure 2c: Landing Phase CCI RF/G

Figure I-11: Graphed means and variances for landing phase co-contraction index variables for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

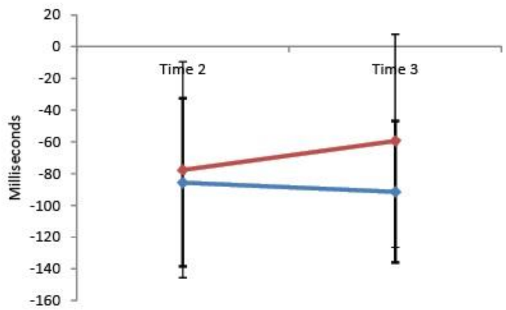


Figure 1a: Rectus Onset

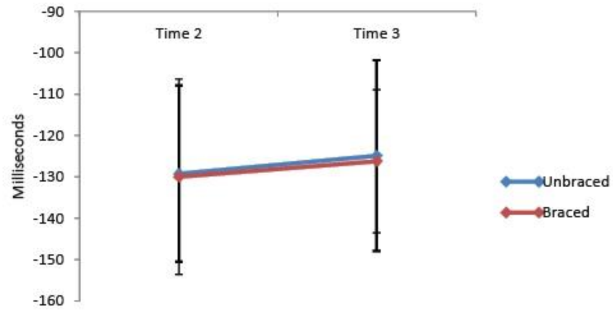


Figure 2a: Biceps Femoris Onset

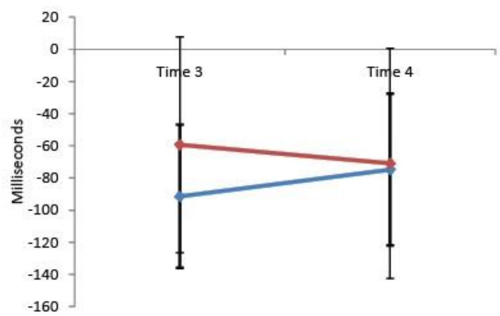


Figure 1b: Rectus Onset

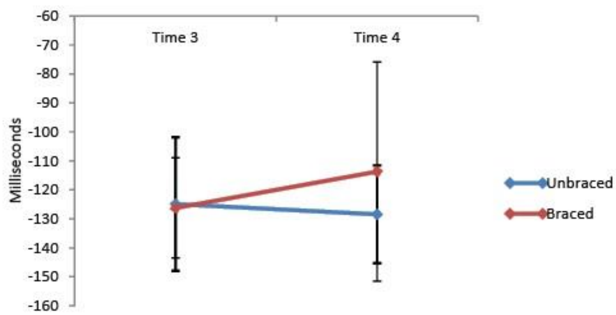


Figure 2b: Biceps Femoris Onset

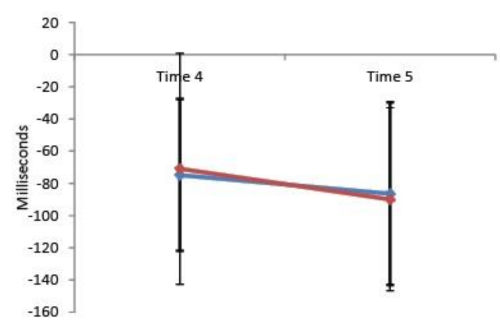


Figure 1c: Rectus Onset

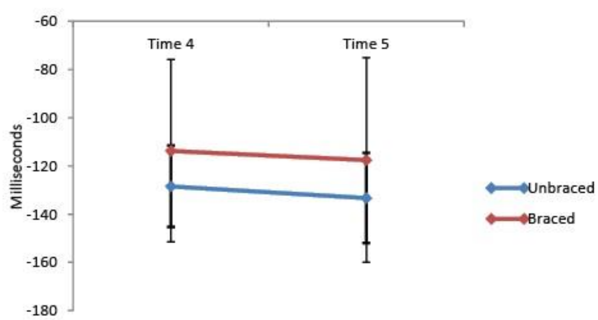


Figure 2c: Biceps Femoris Onset

Figure I-12: Graphed means and variances for rectus femoris and biceps femoris onset for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

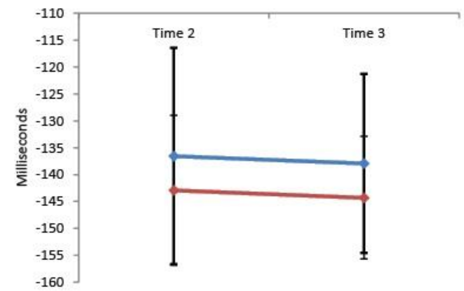


Figure 1a: Gastrocnemius Onset

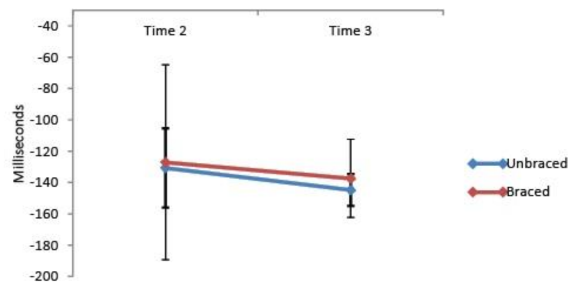


Figure 2a: Gluteus Medius Onset

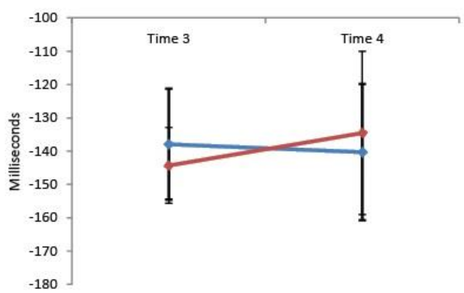


Figure 1b: Gastrocnemius Onset

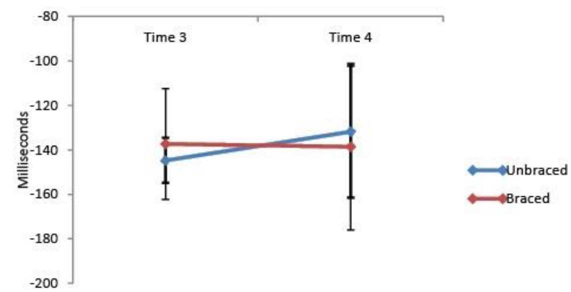


Figure 2b: Gluteus Medius Onset

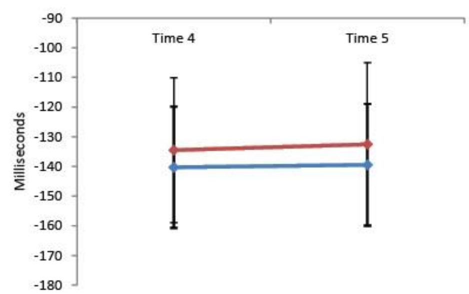


Figure 1c: Gastrocnemius Onset

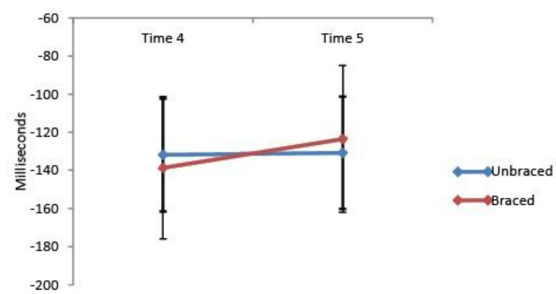


Figure 2c: Gluteus Medius Onset

Figure I-13: Graphed means and variances for gastrocnemius and gluteus medius onset for each hypothesis. A) Hypothesis 1 between times 2 and 3; B) Hypothesis 2 between times 3 and 4; and C) Hypothesis 3 between times 4 and 5.

Appendix J: Visual3D Joint Angle Reports - Means and Standard Deviations

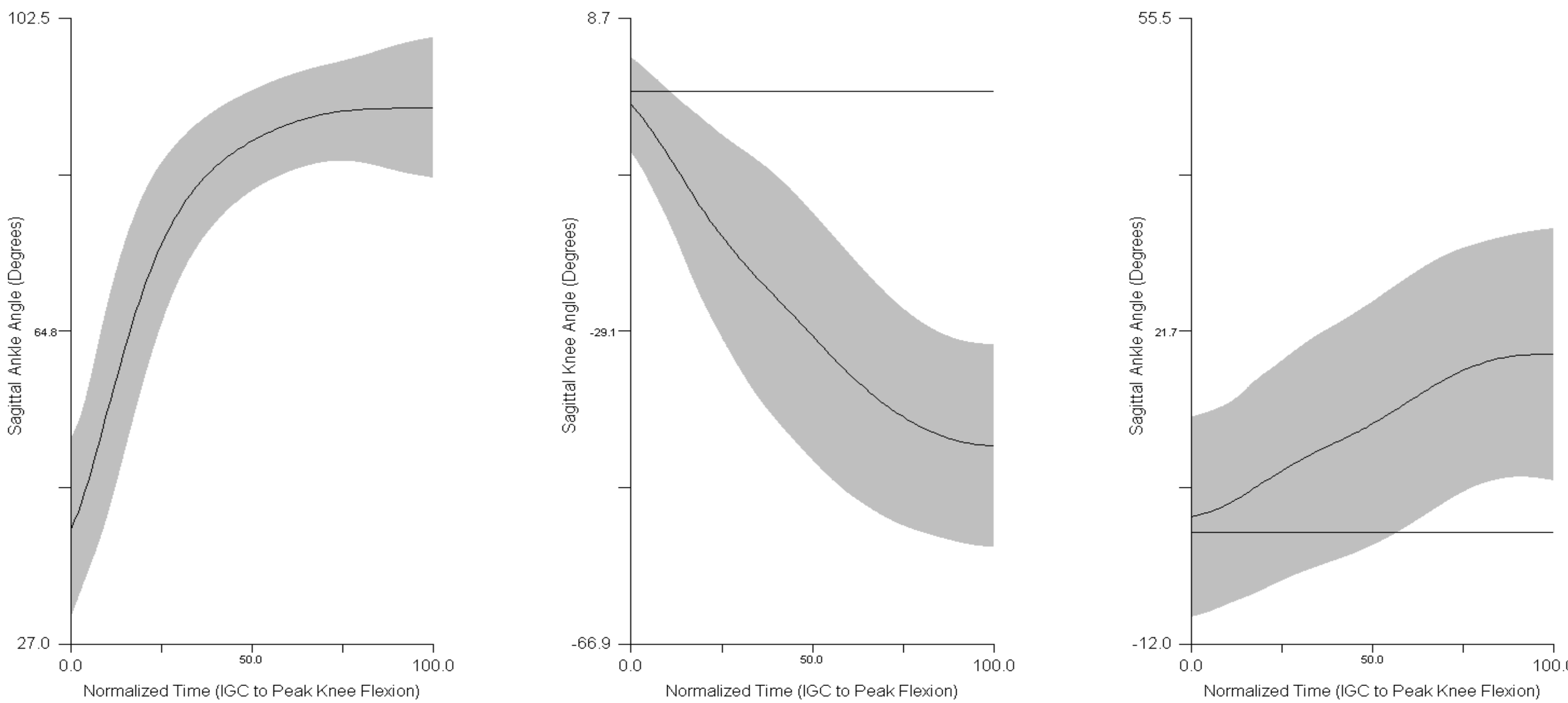


Figure J-1: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

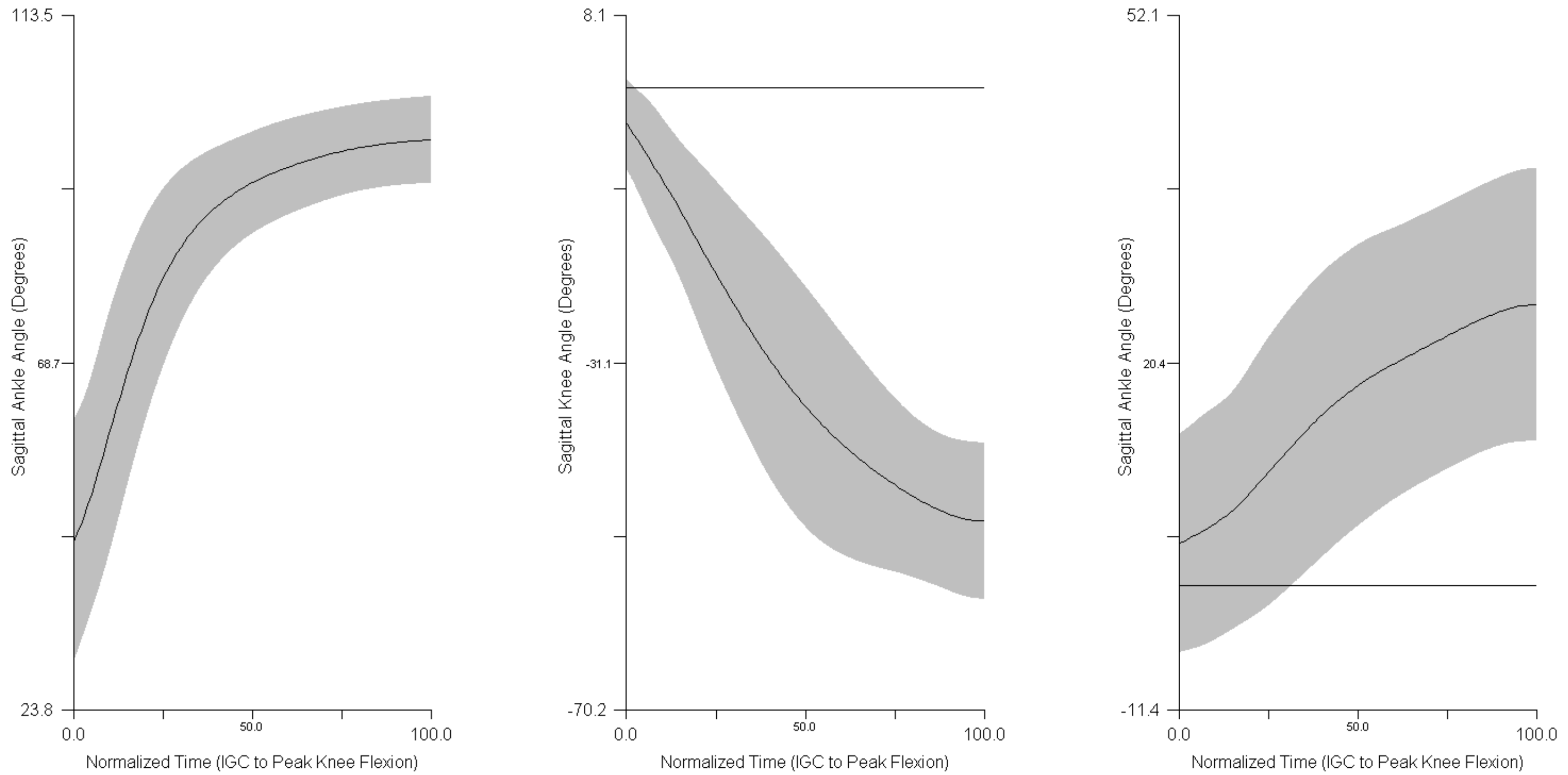


Figure J-2: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **braced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

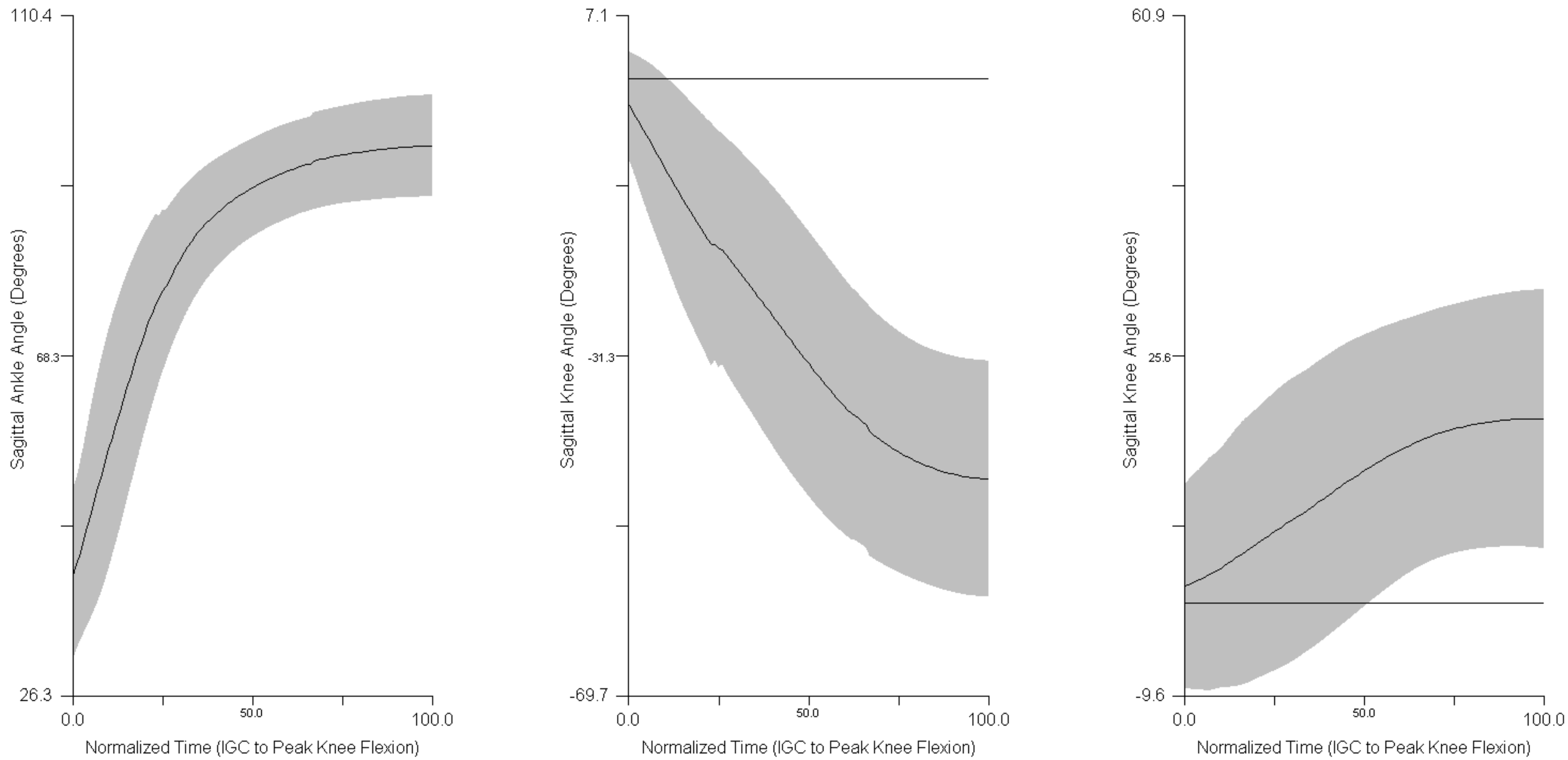


Figure J-3: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

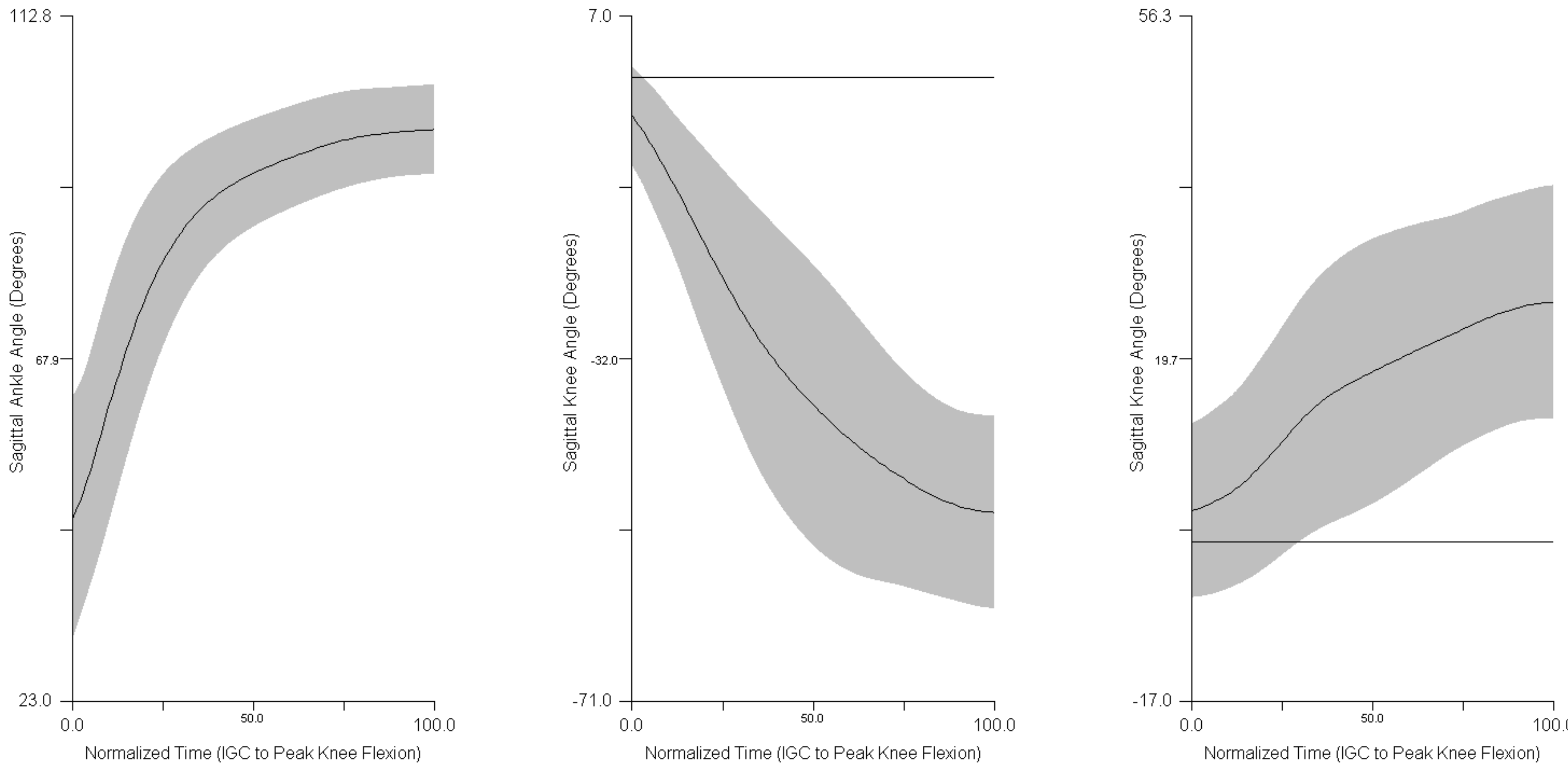


Figure J-4: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **braced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

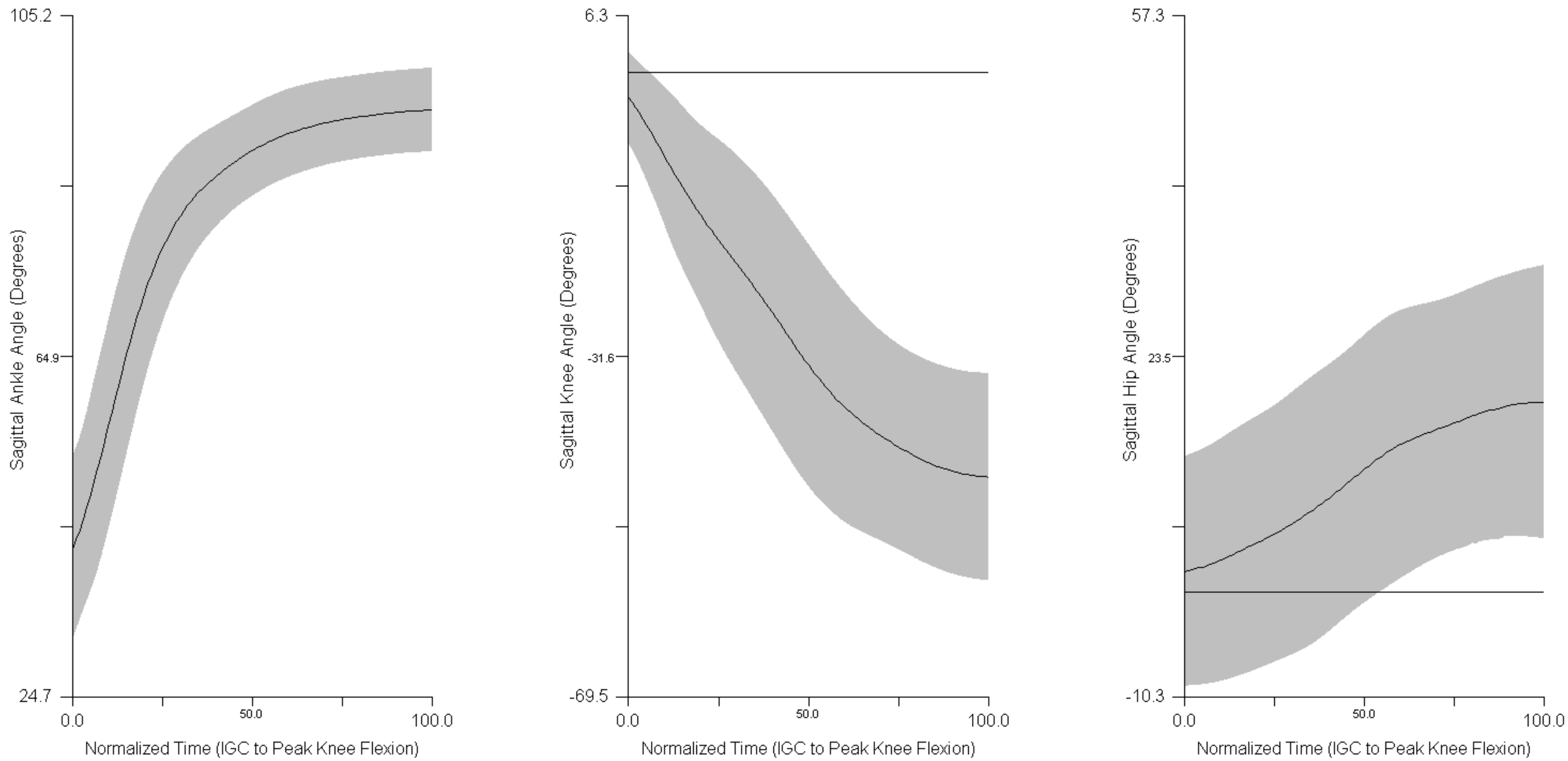


Figure J-5: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

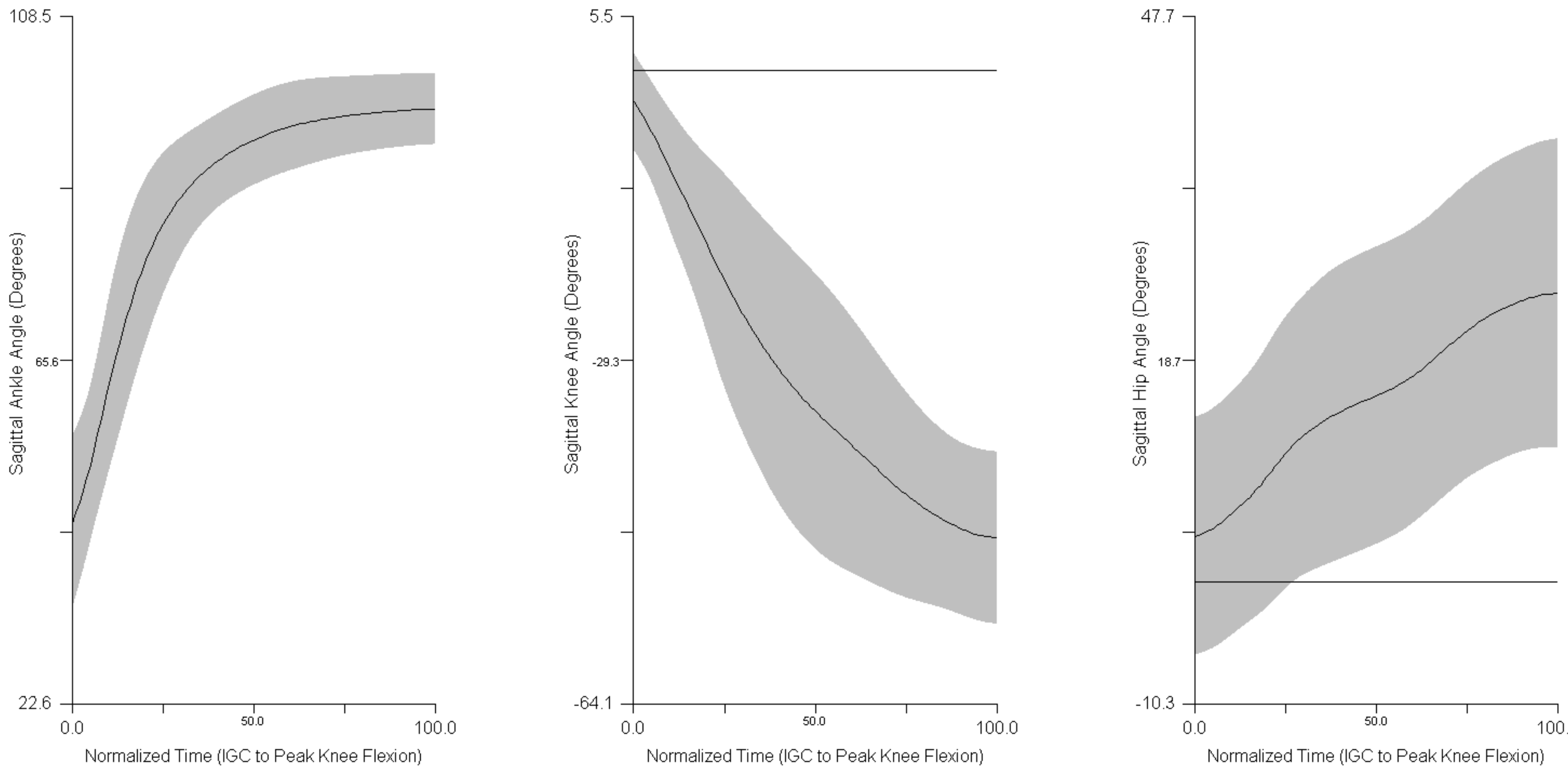


Figure J-6: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **braced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

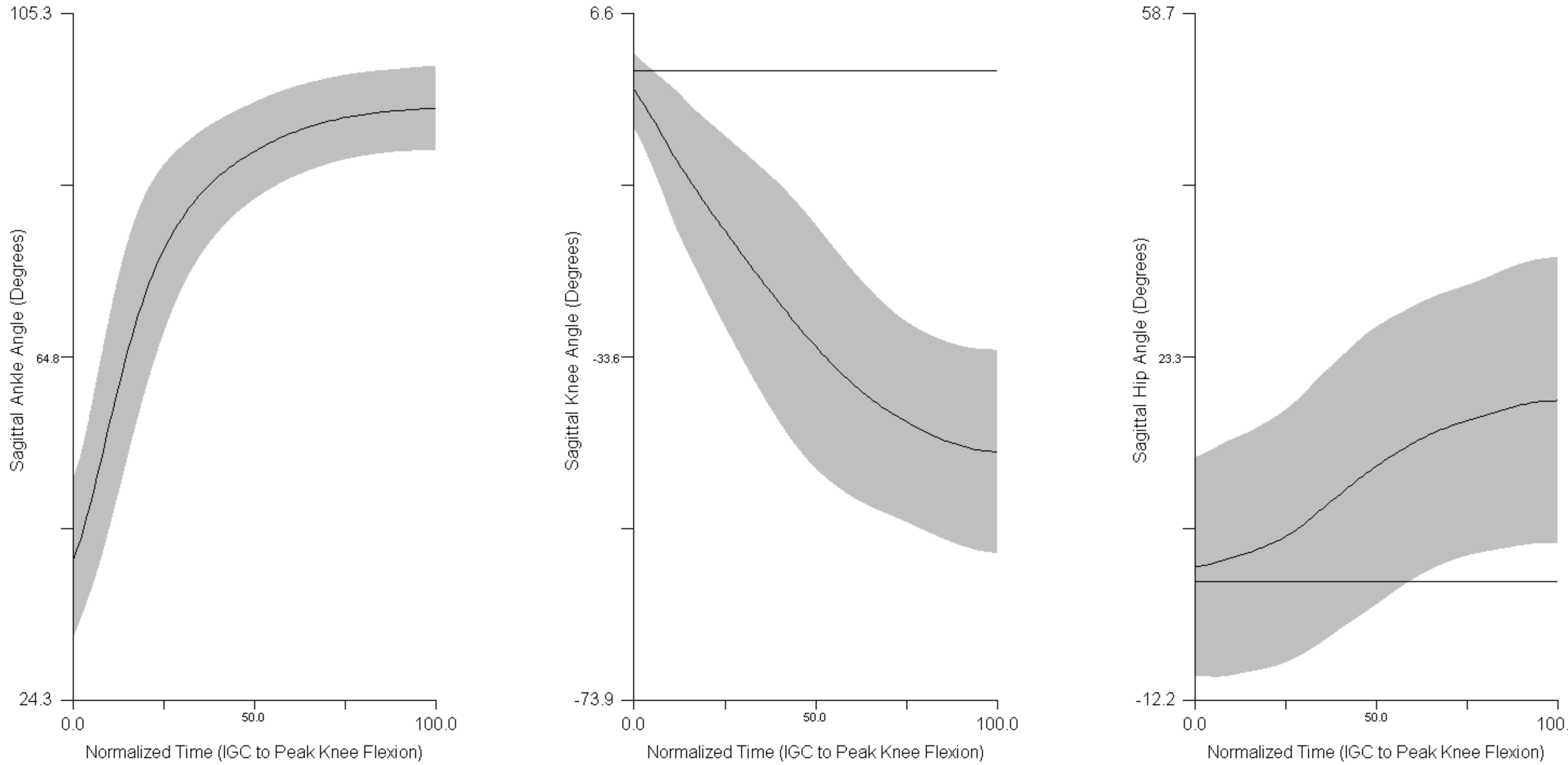


Figure J-7: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

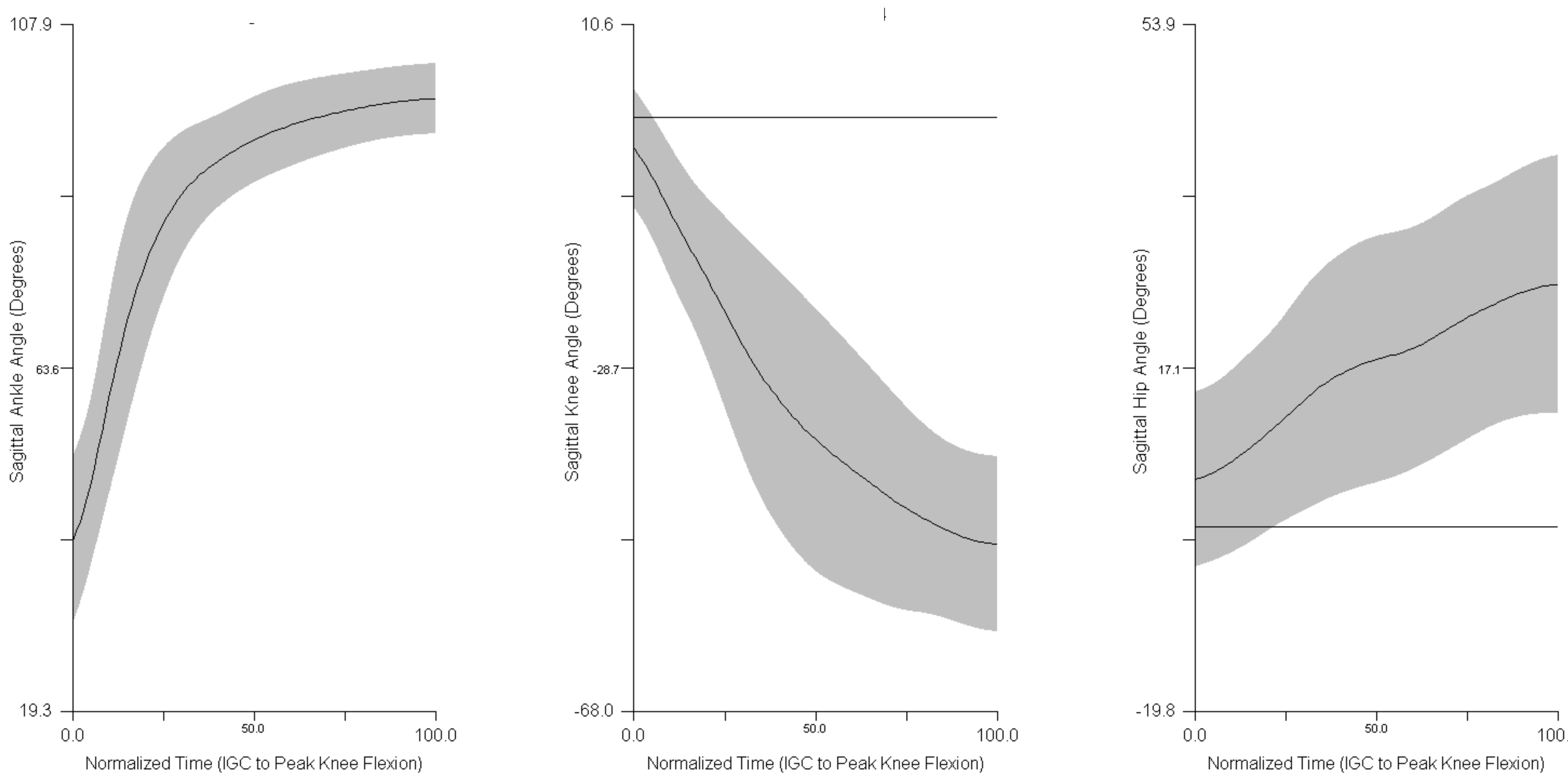


Figure J-8: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **braced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

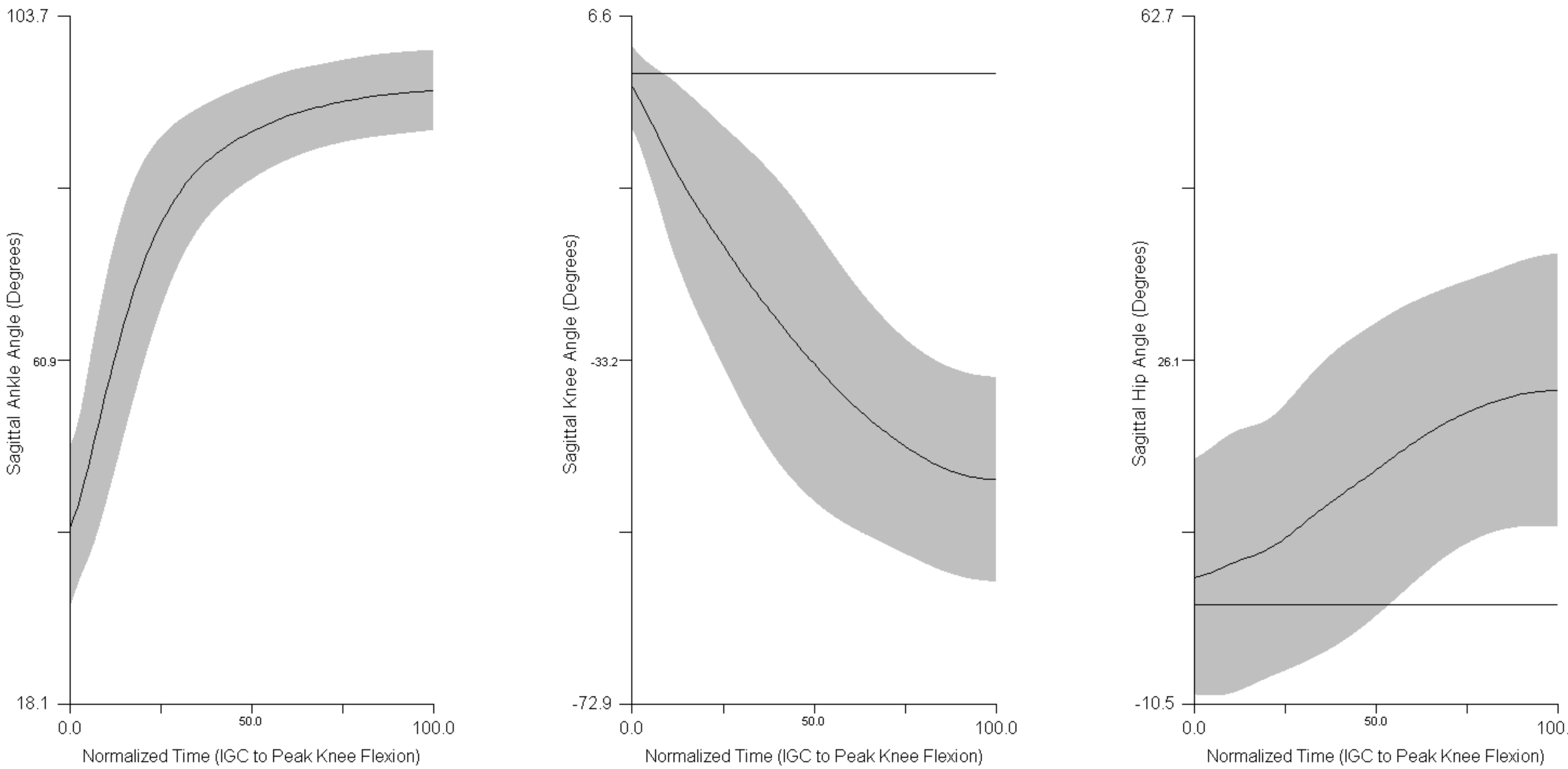


Figure J-9: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

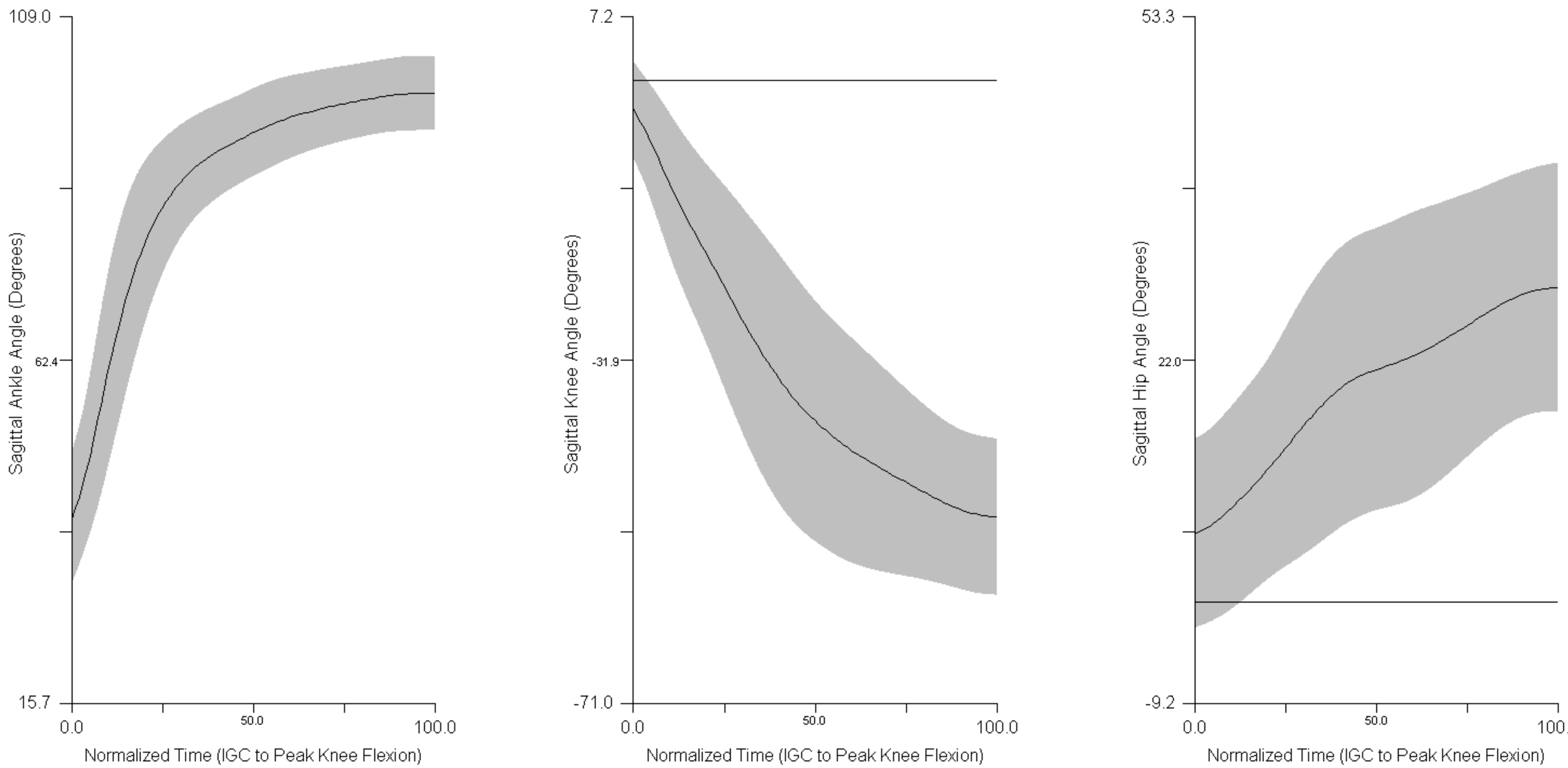


Figure J-10: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) **braced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

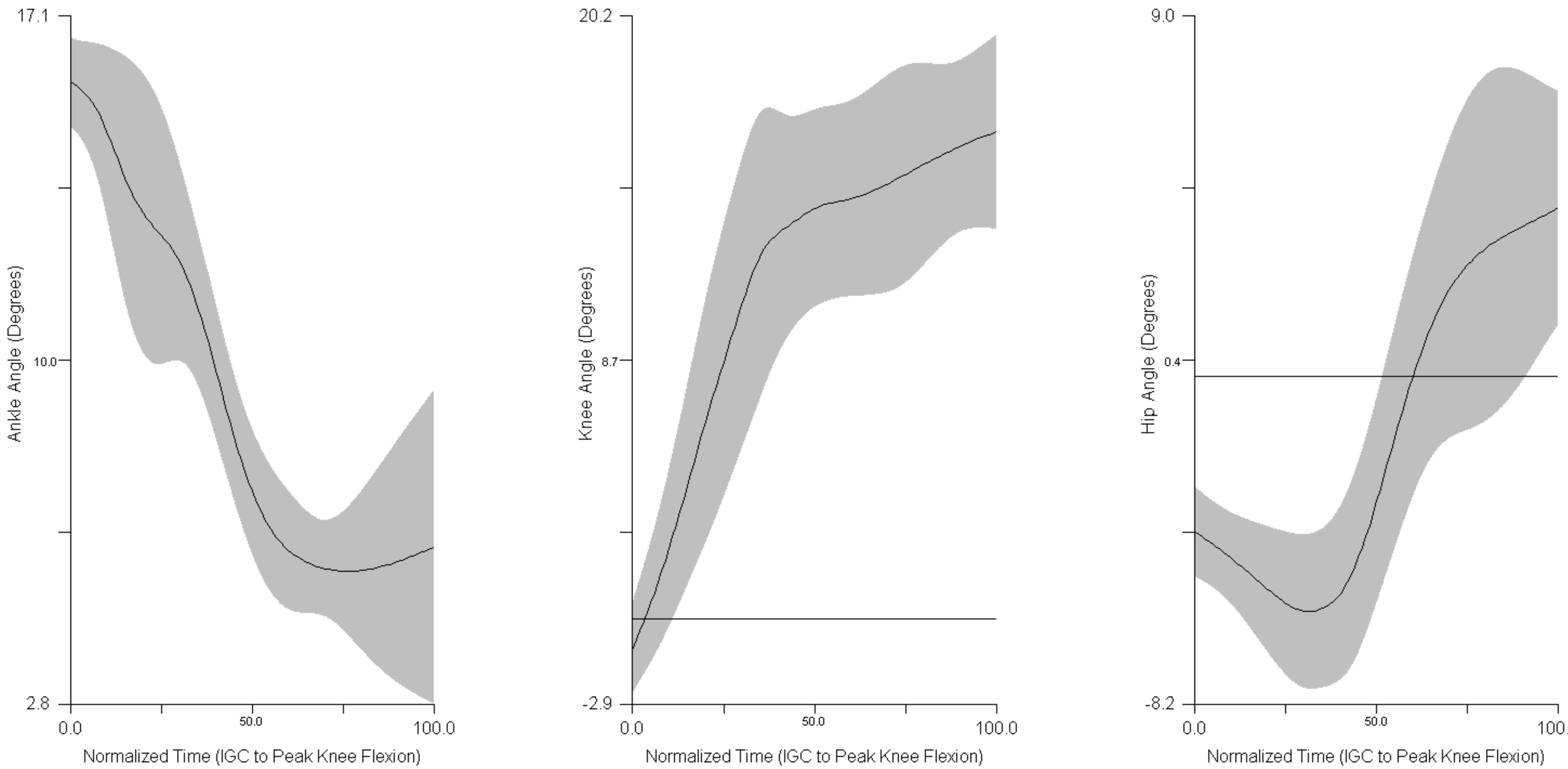


Figure J-11: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

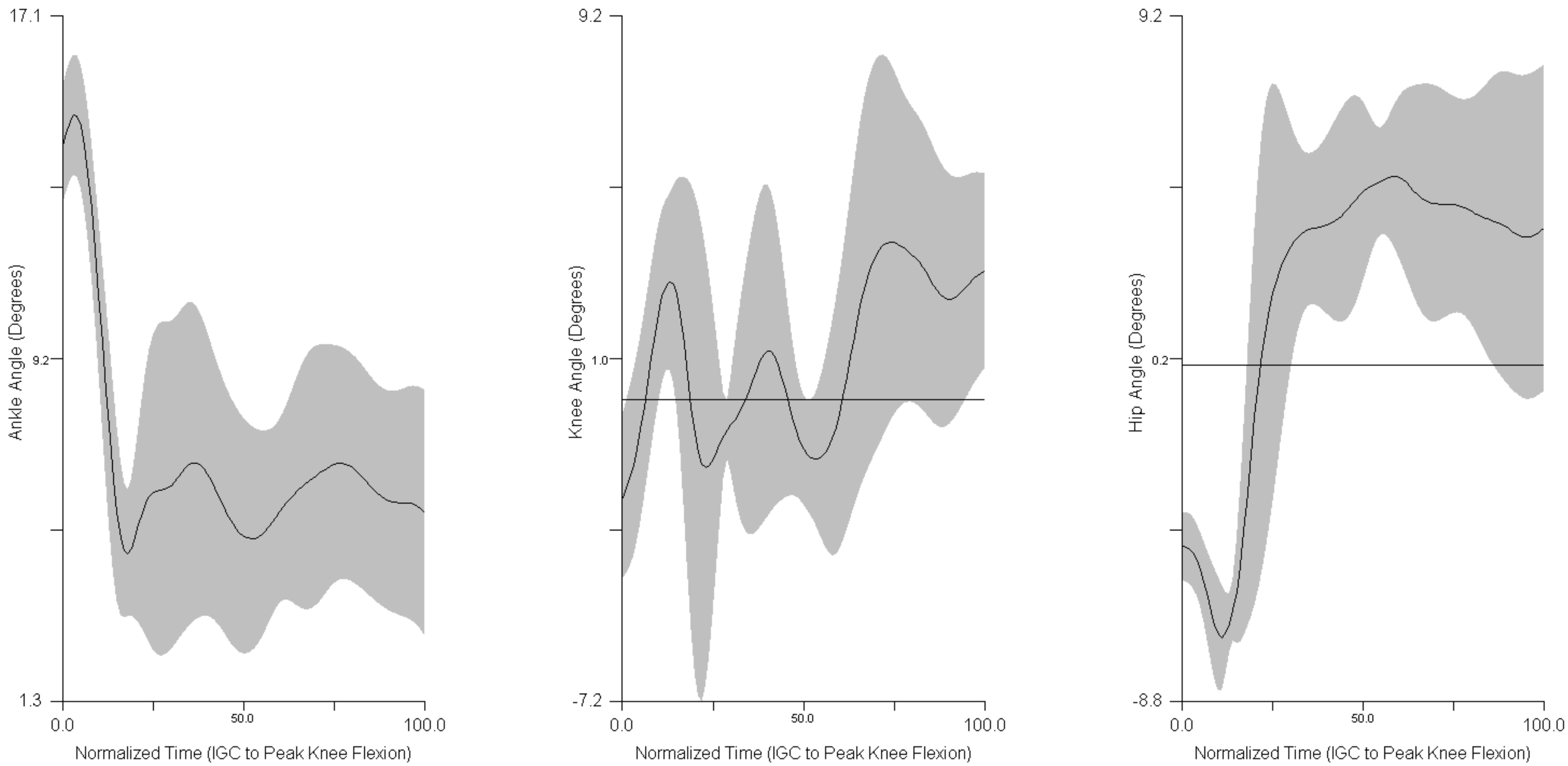


Figure J-12: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **braced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

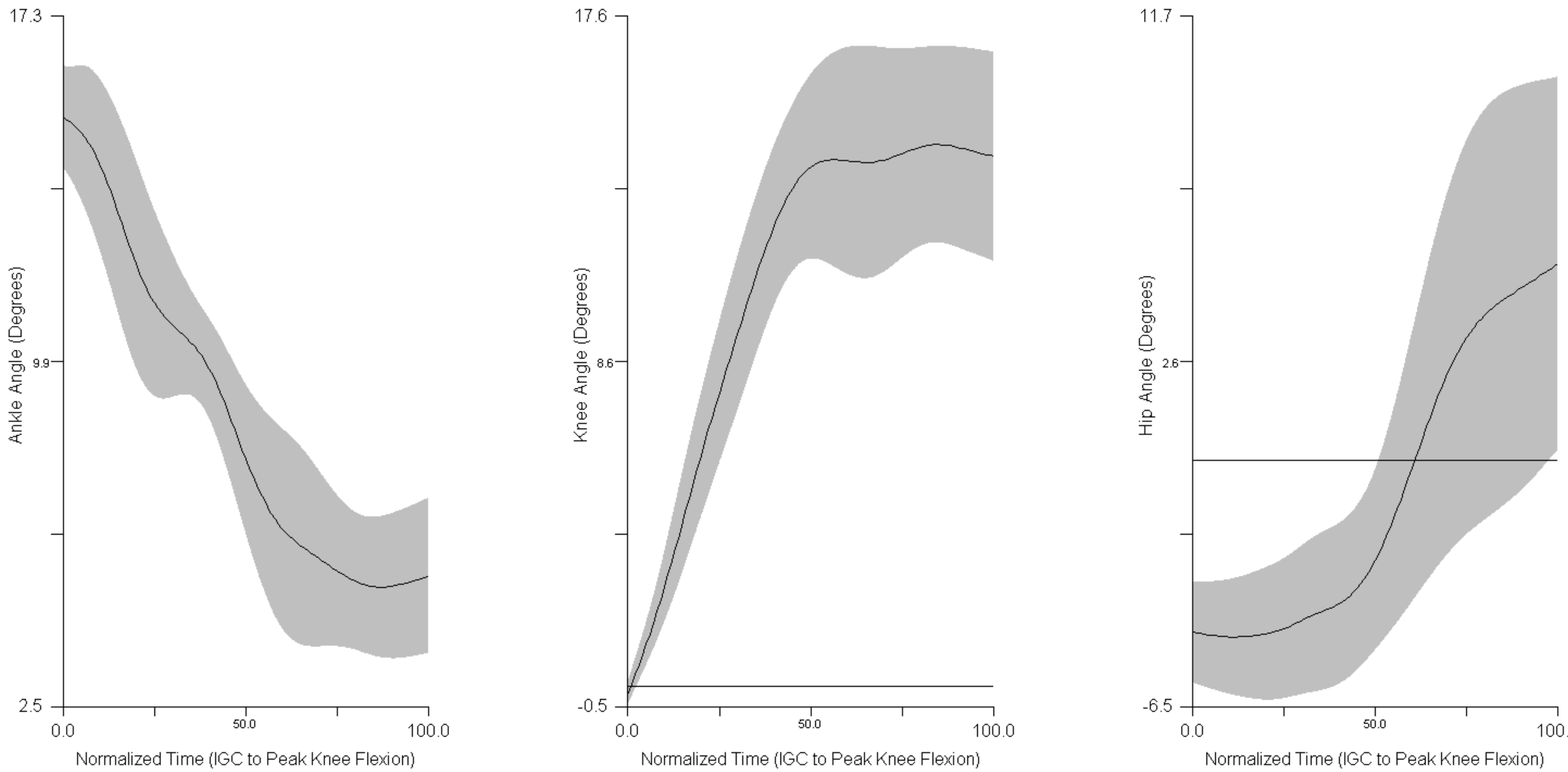


Figure J-13: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

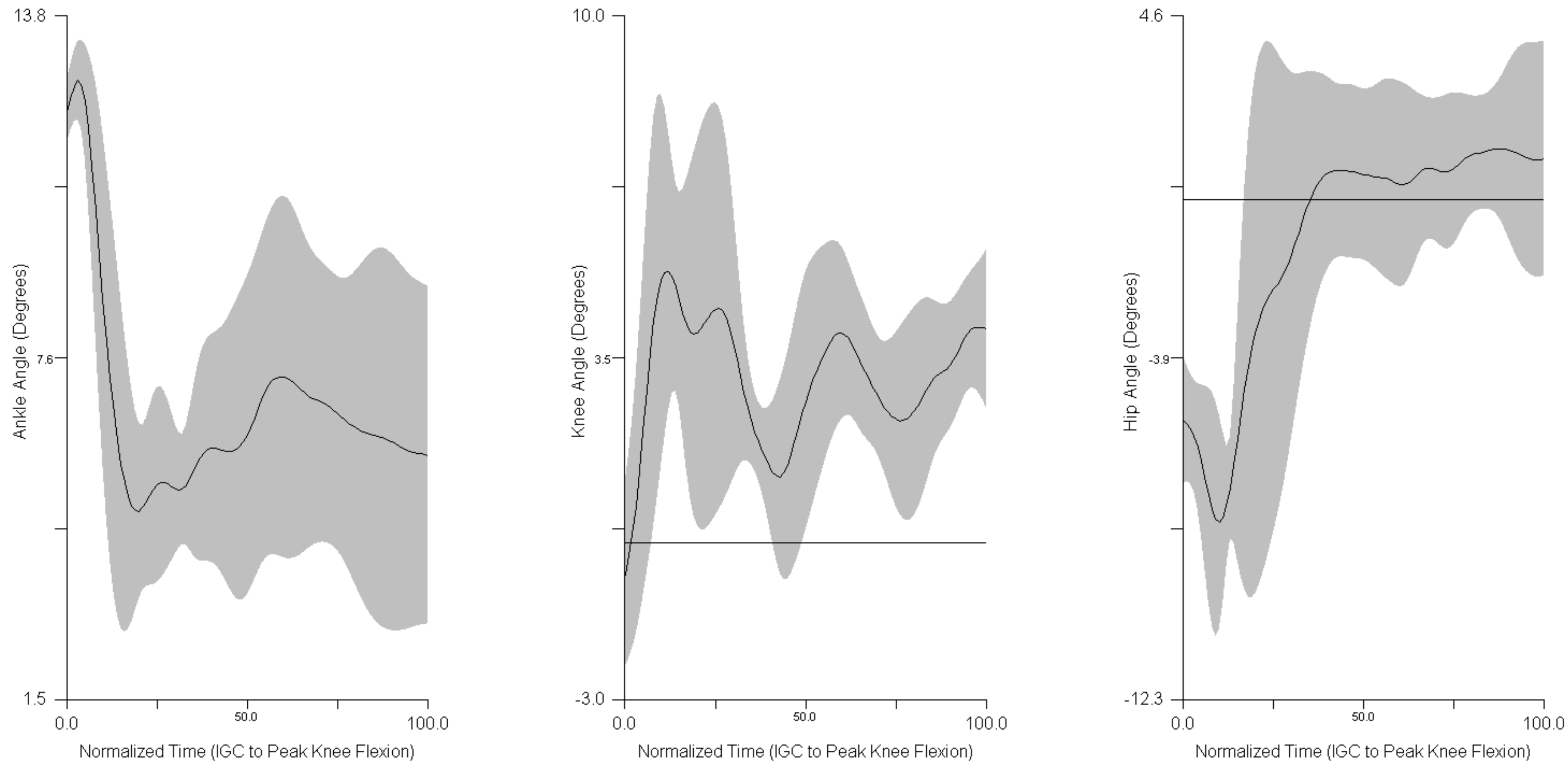


Figure J-14: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **braced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

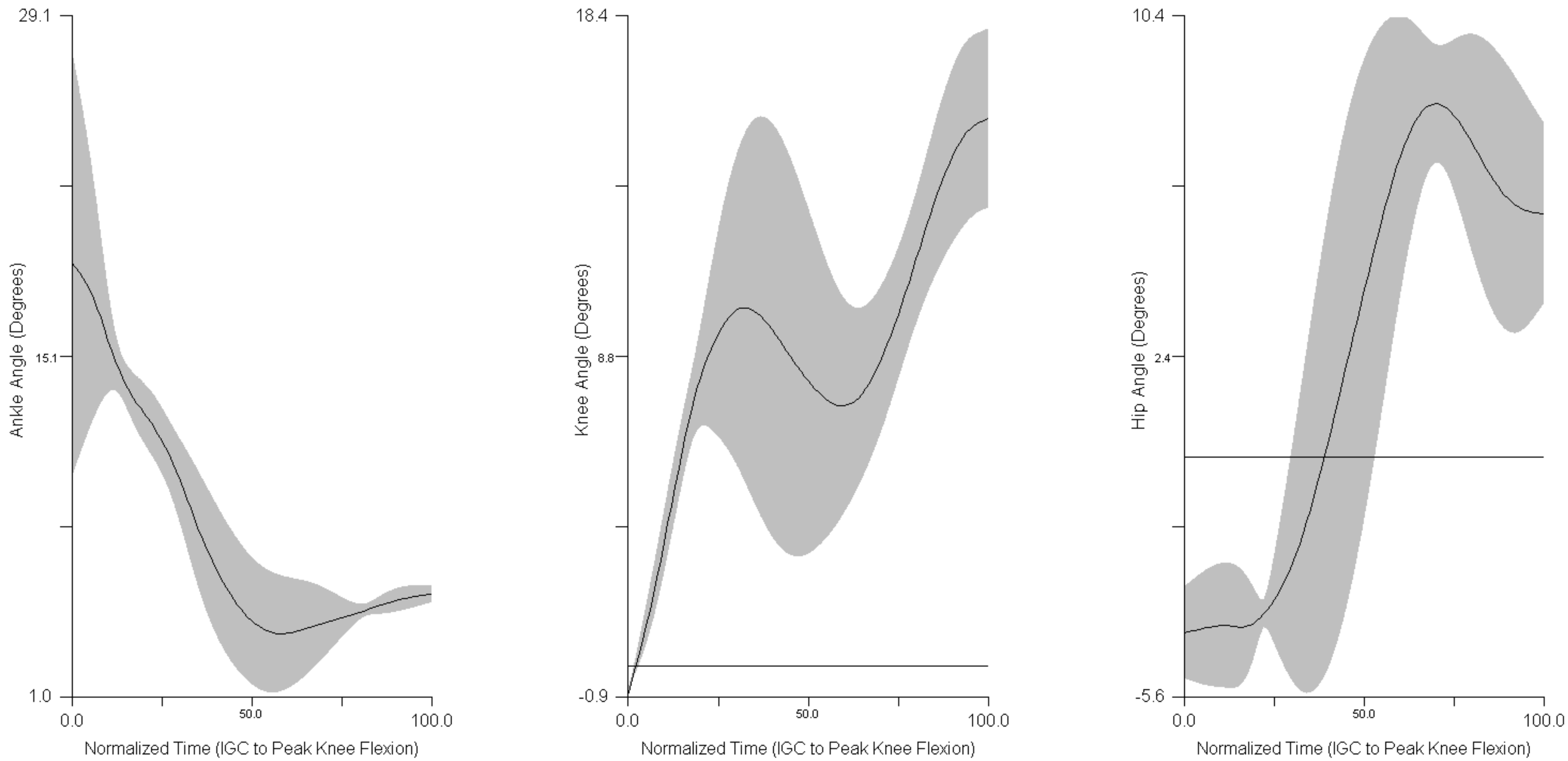


Figure J-15: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

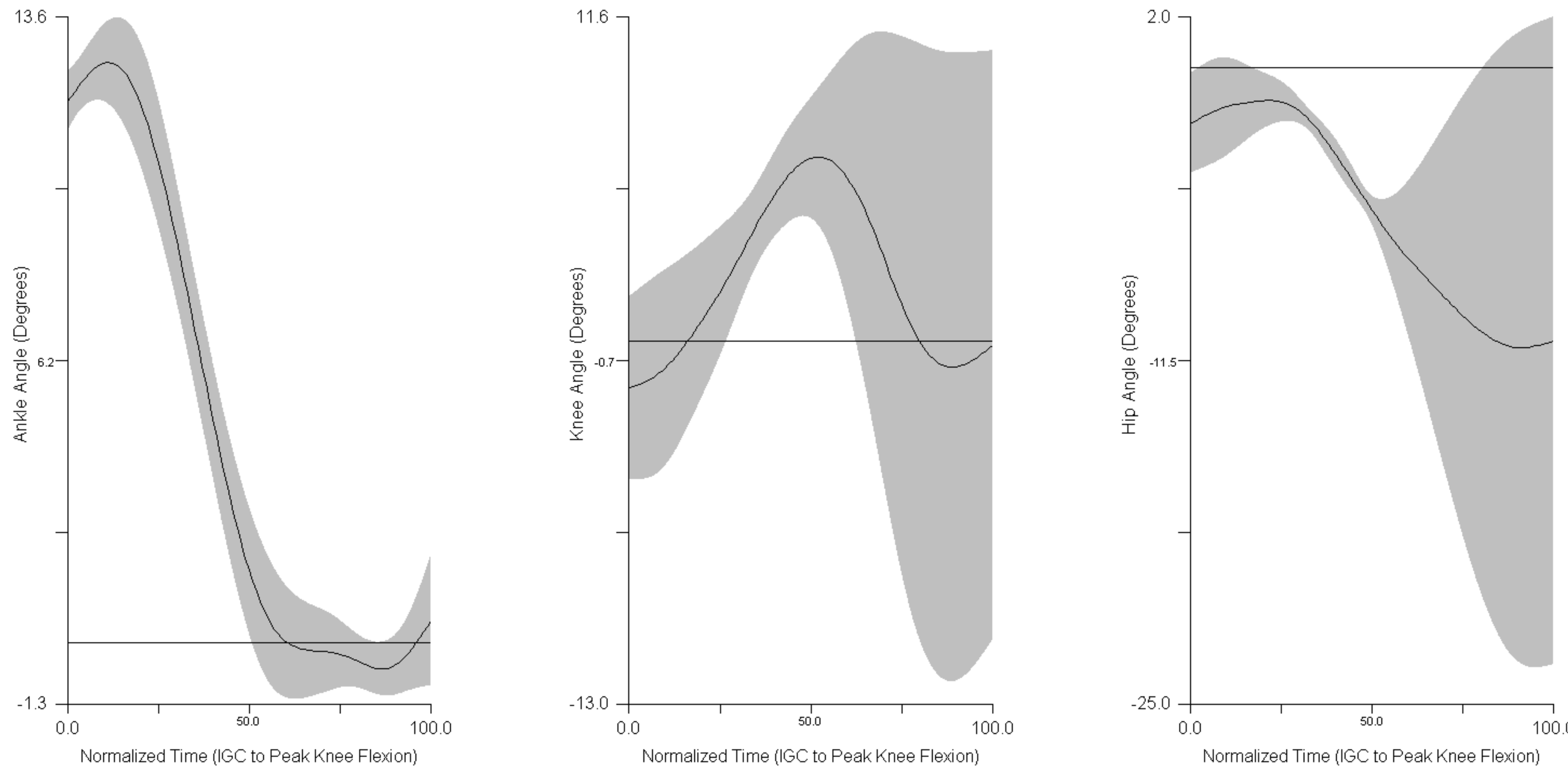


Figure J-16: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **braced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

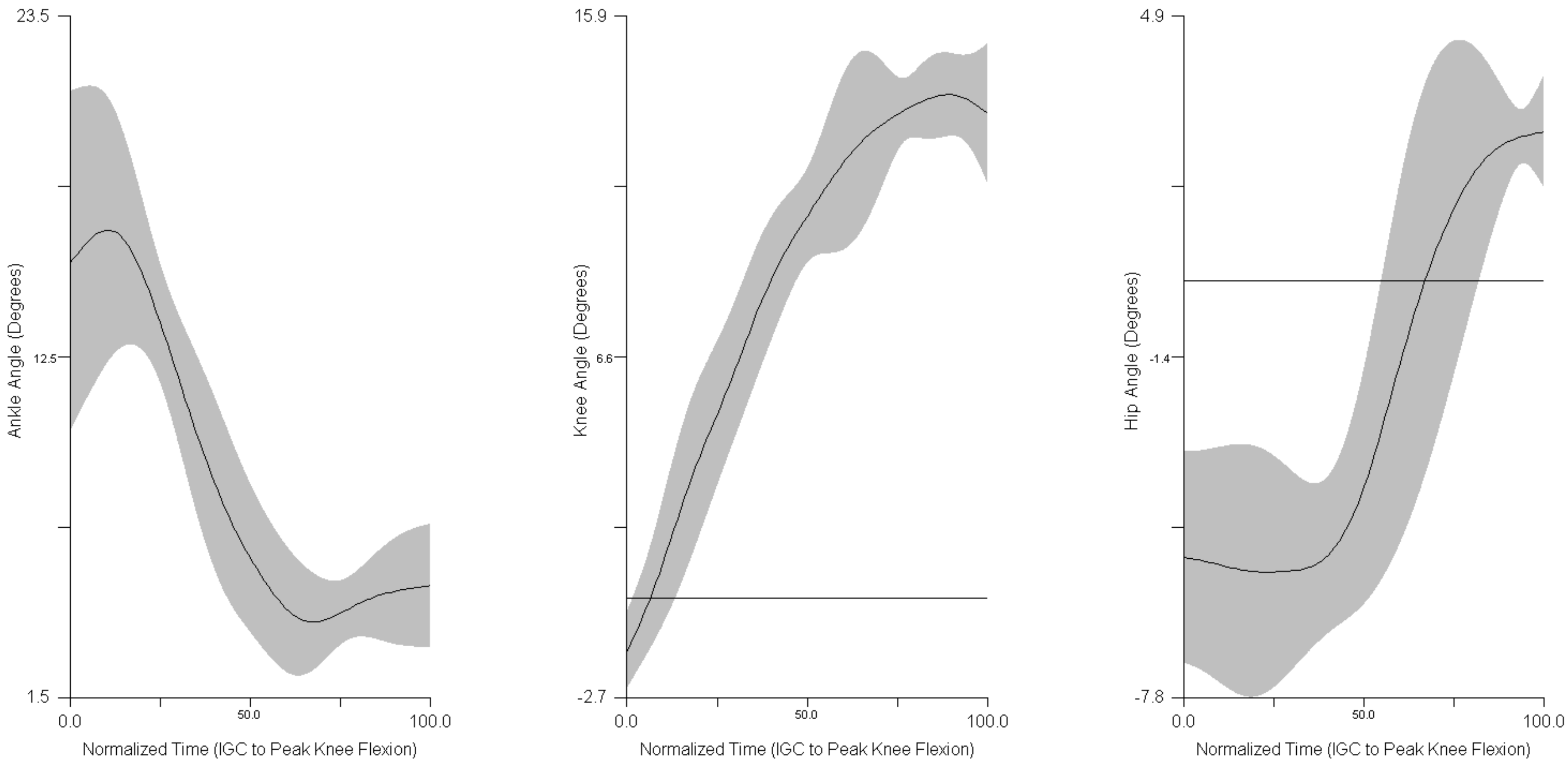


Figure J-17: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

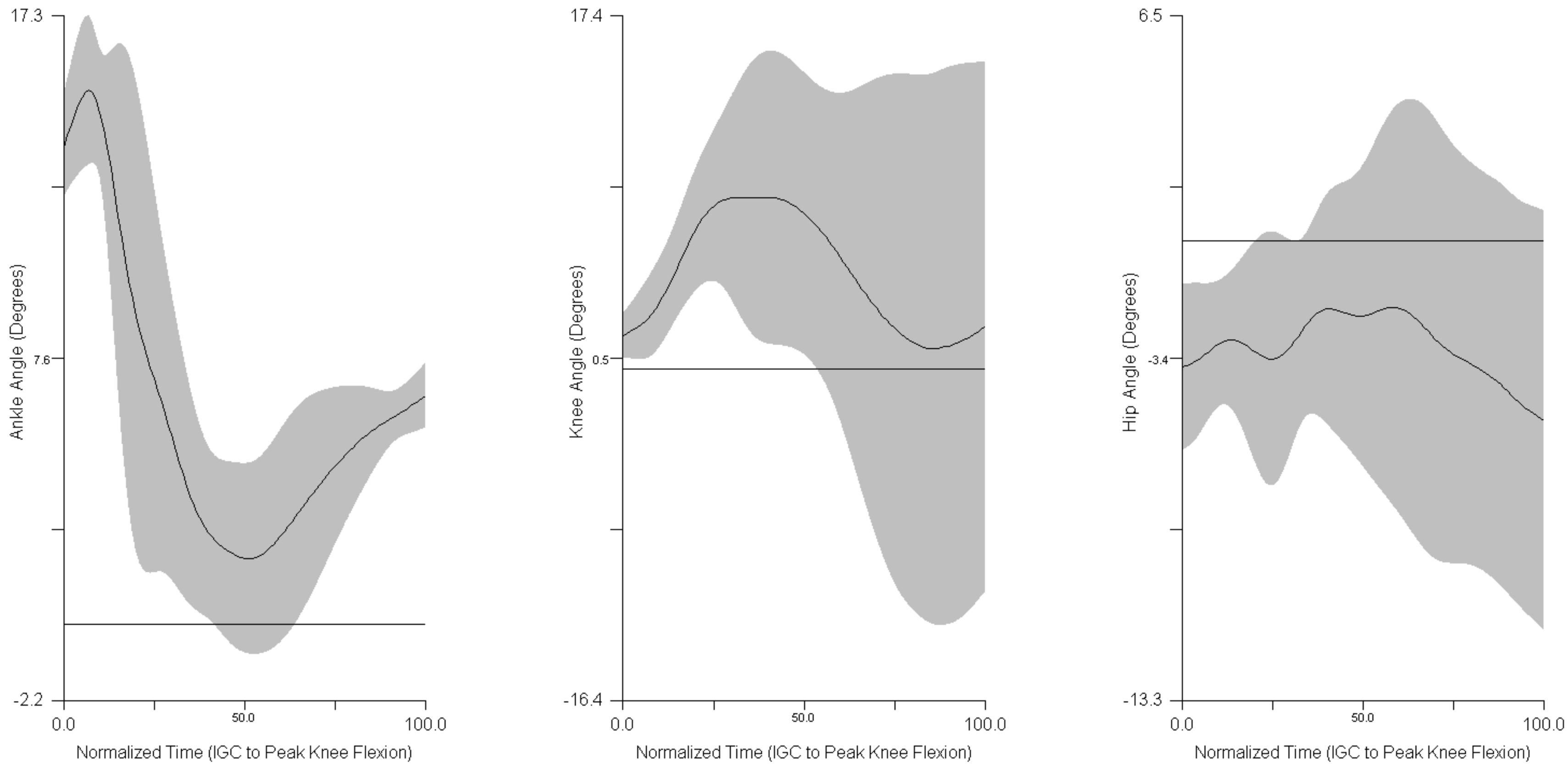


Figure J-18: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **braced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

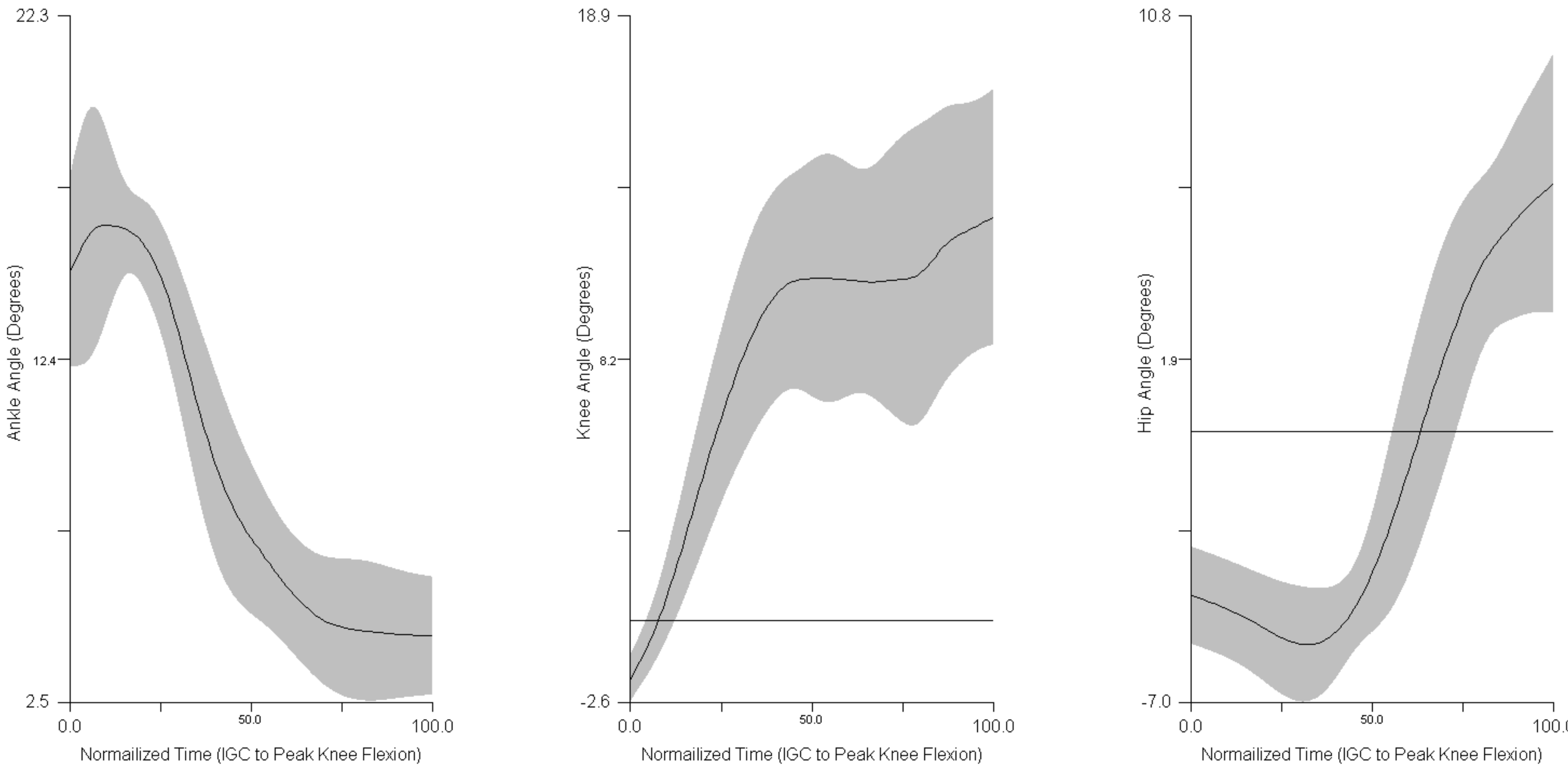


Figure J-19: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **unbraced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

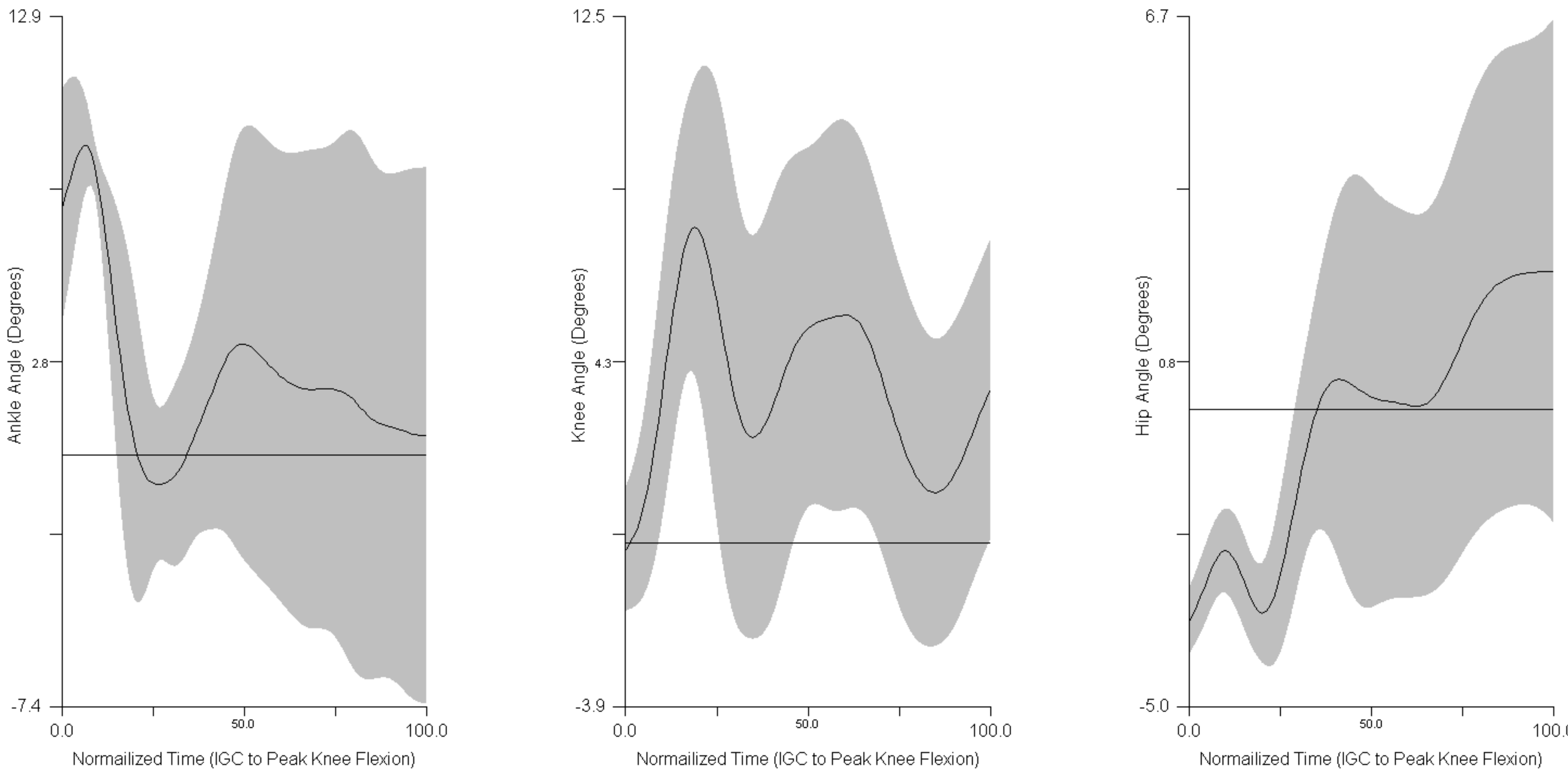


Figure J-20: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) **braced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

Appendix K: Visual3D Joint Moment Reports – Means and Standard Deviations

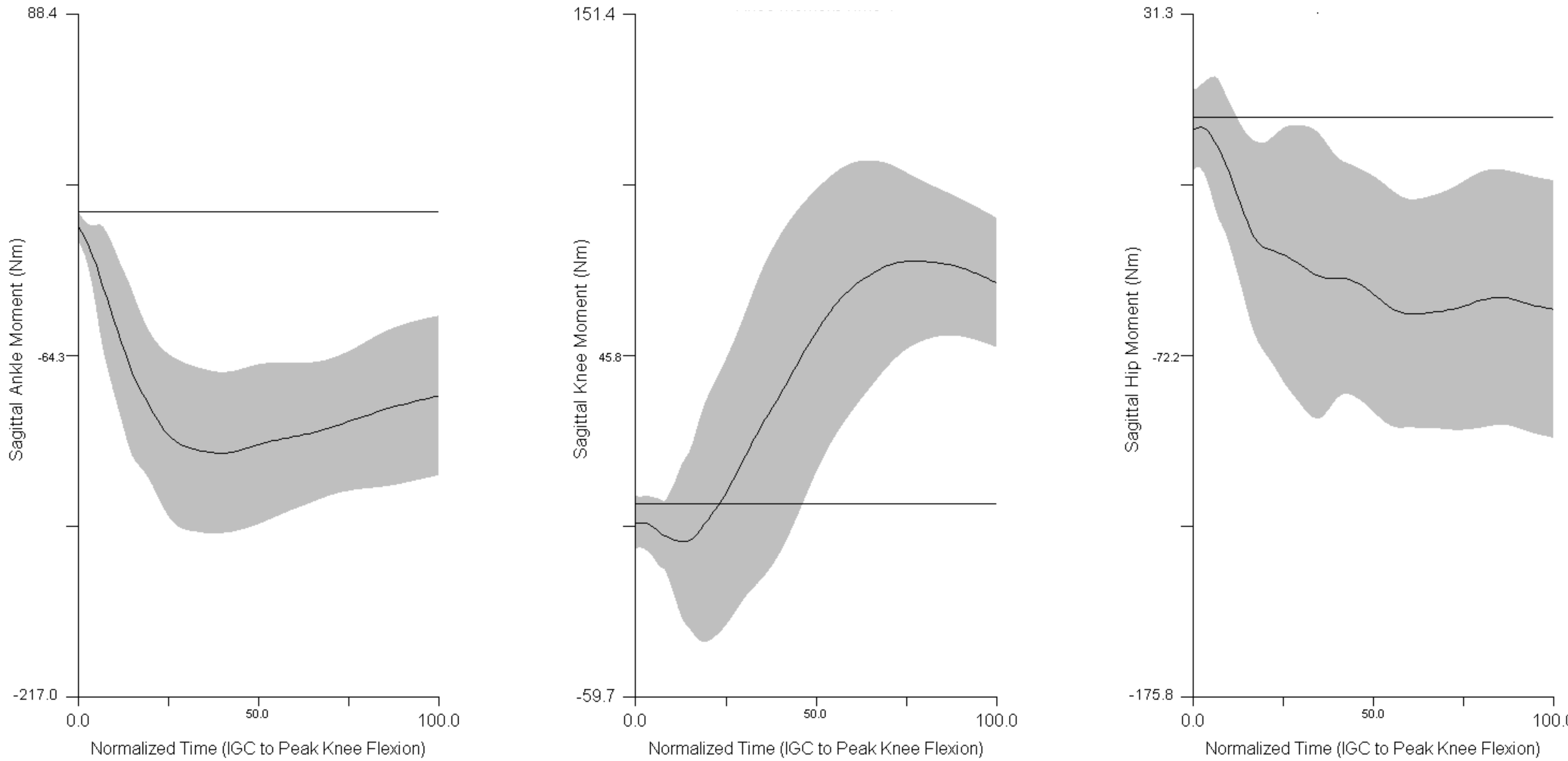


Figure K-1: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

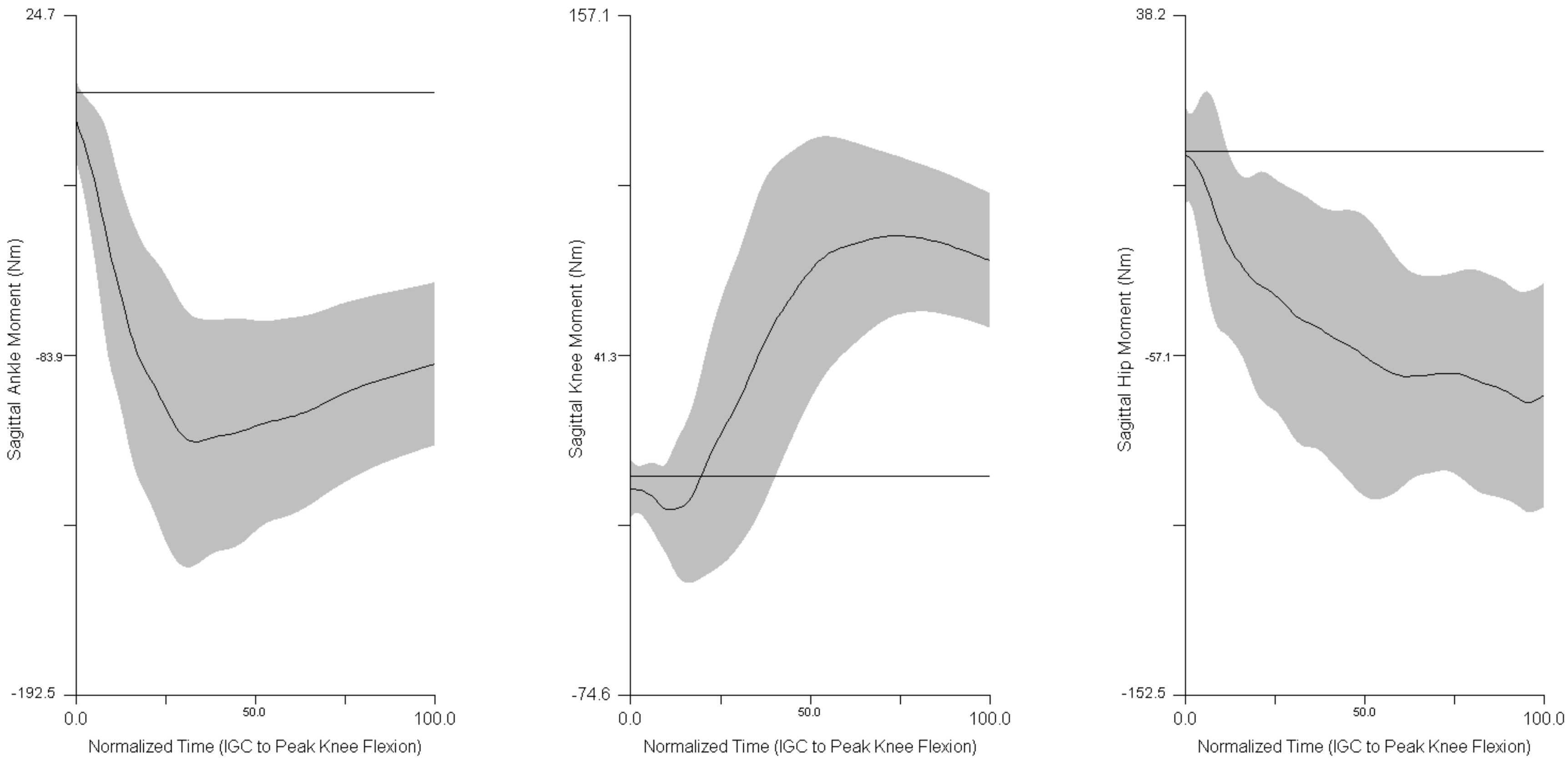


Figure K-2: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

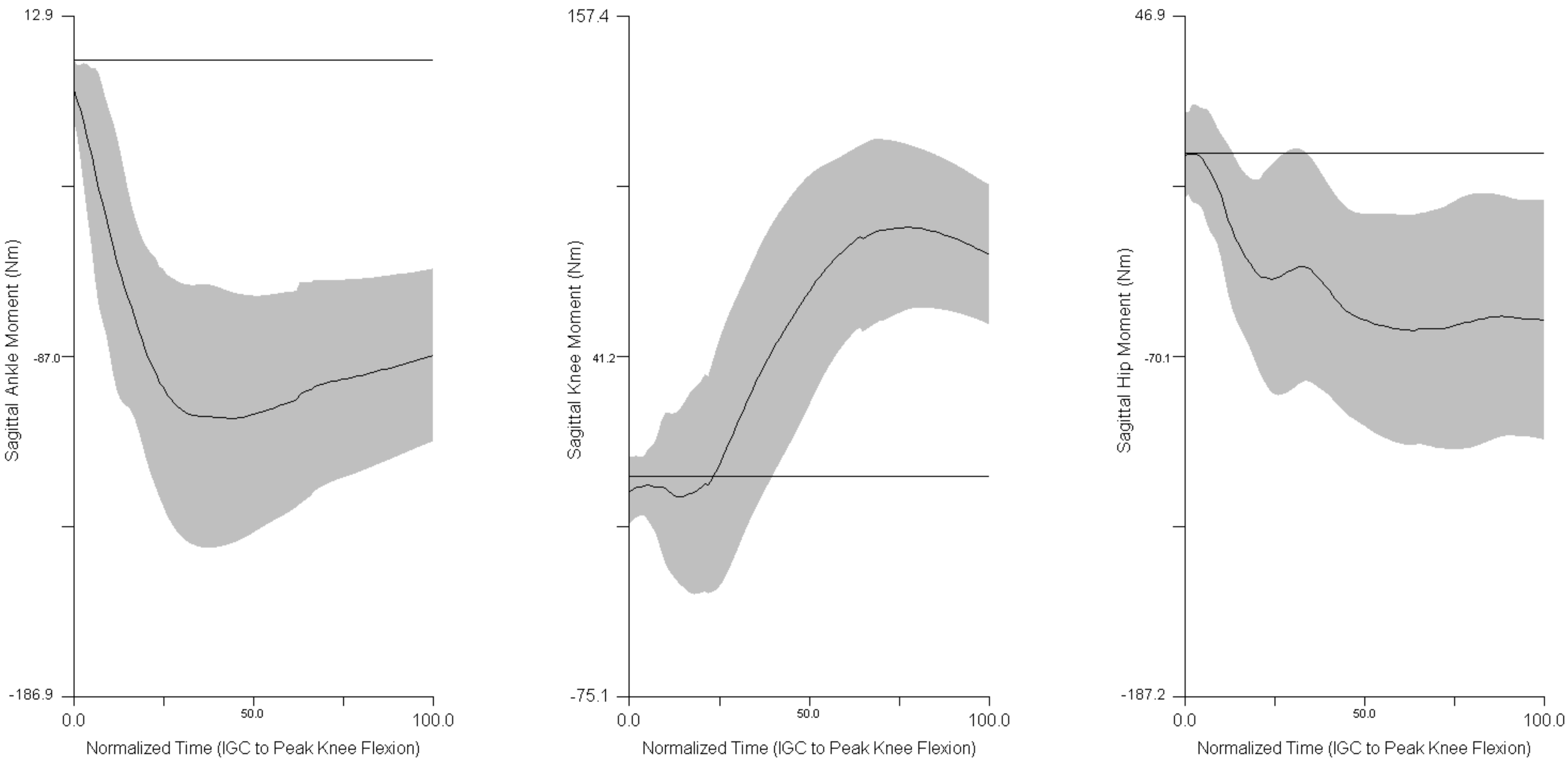


Figure K-3: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

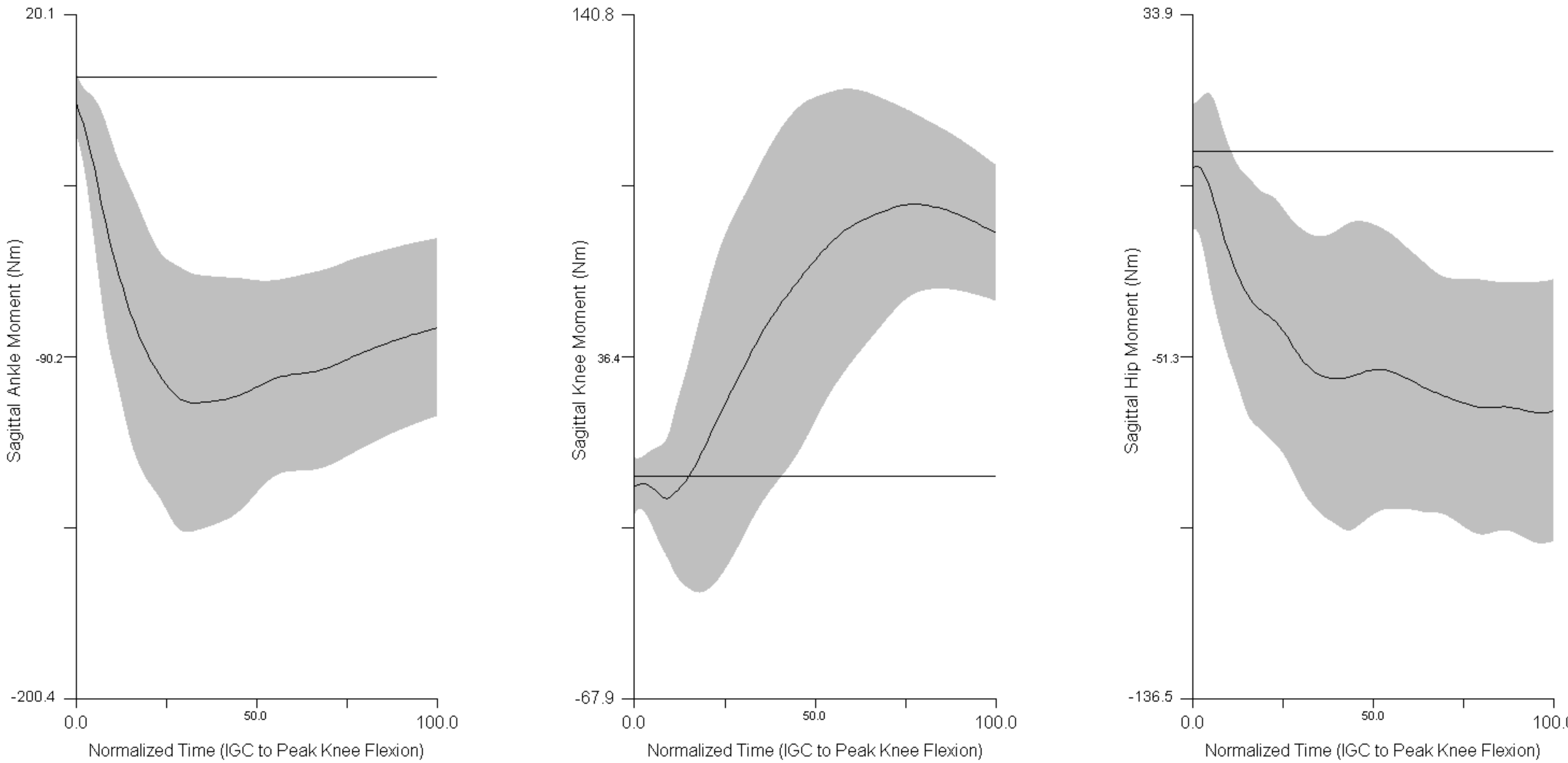


Figure K-4: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

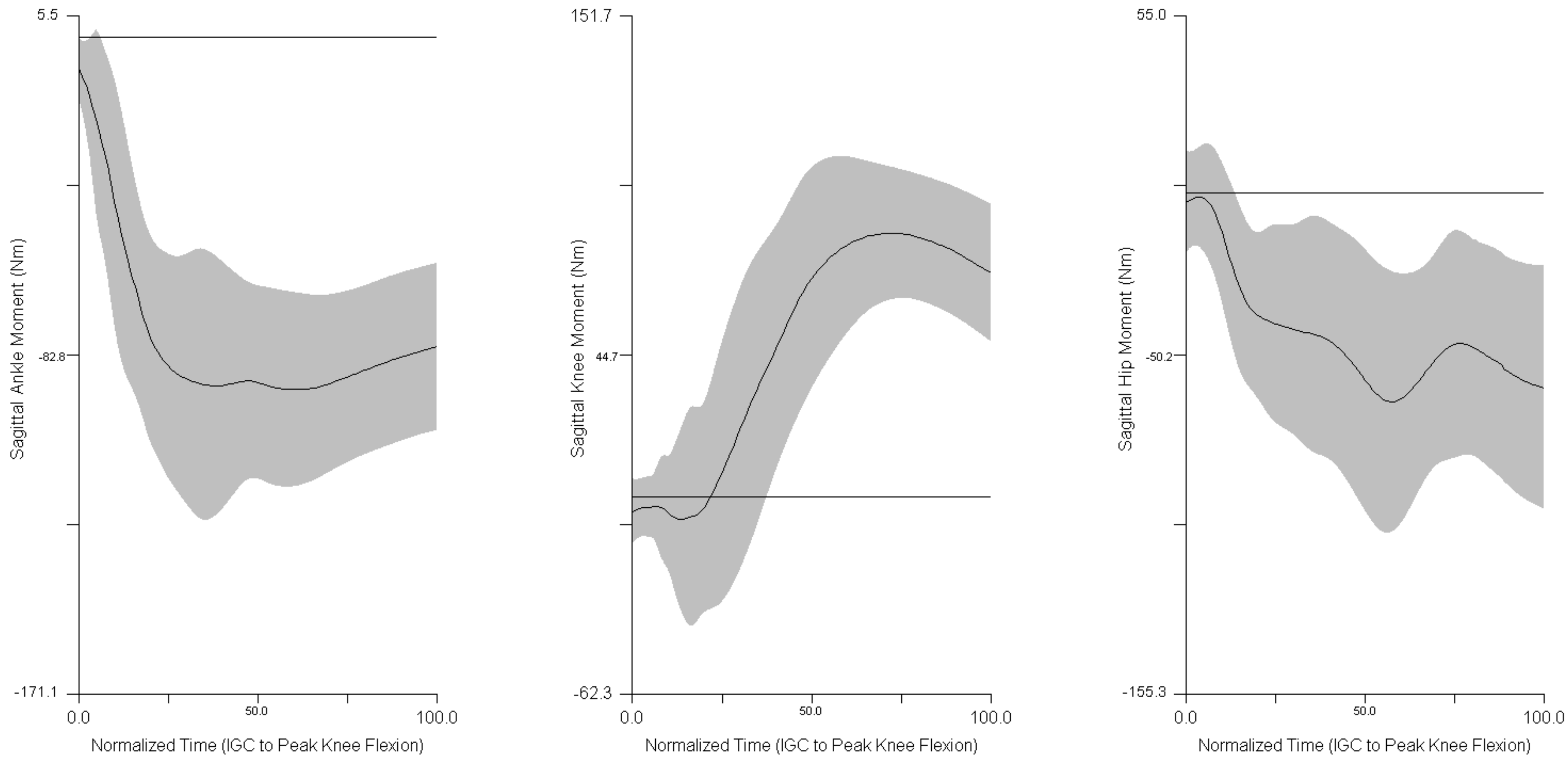


Figure K-5: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

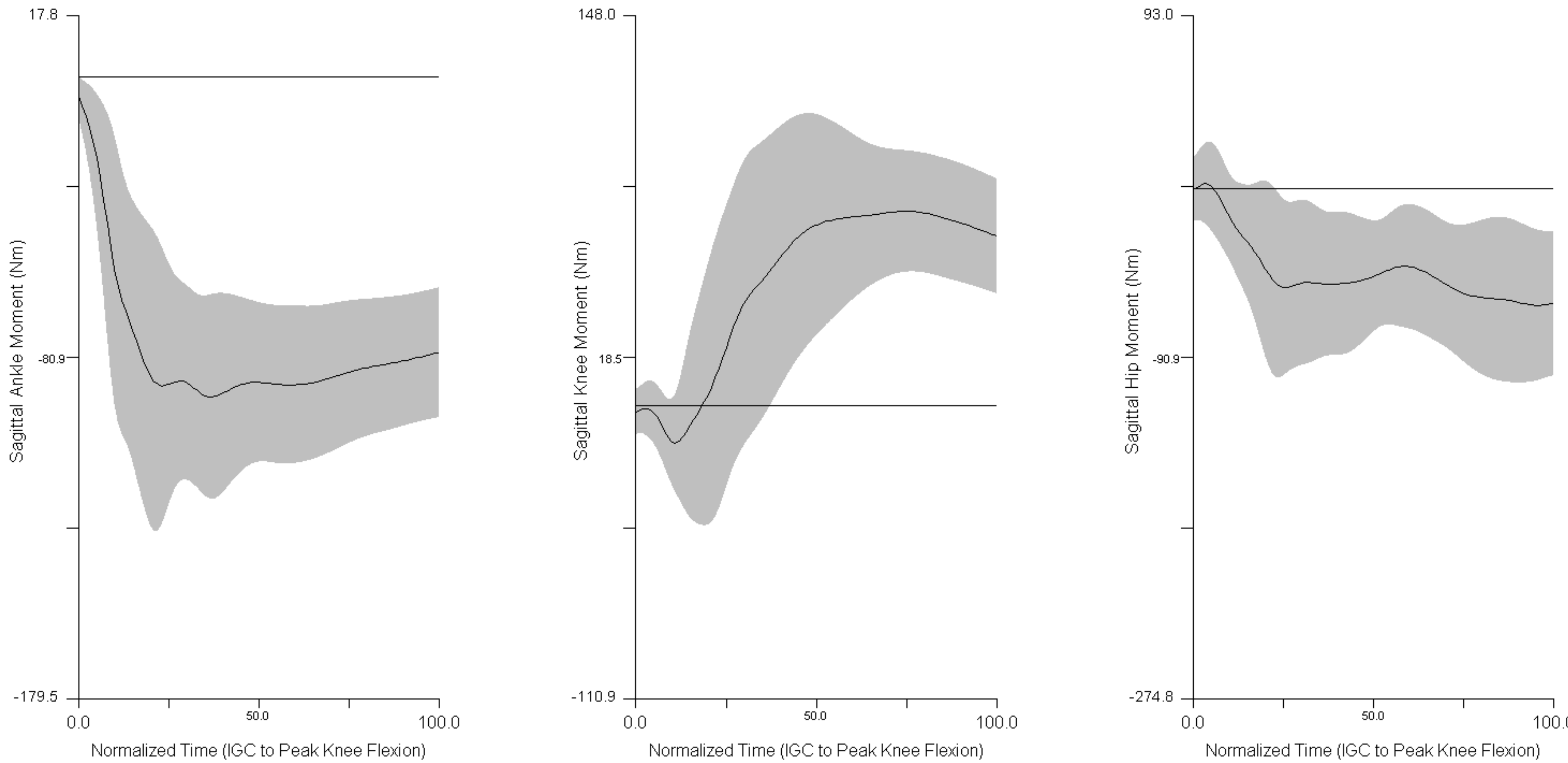


Figure K-6: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

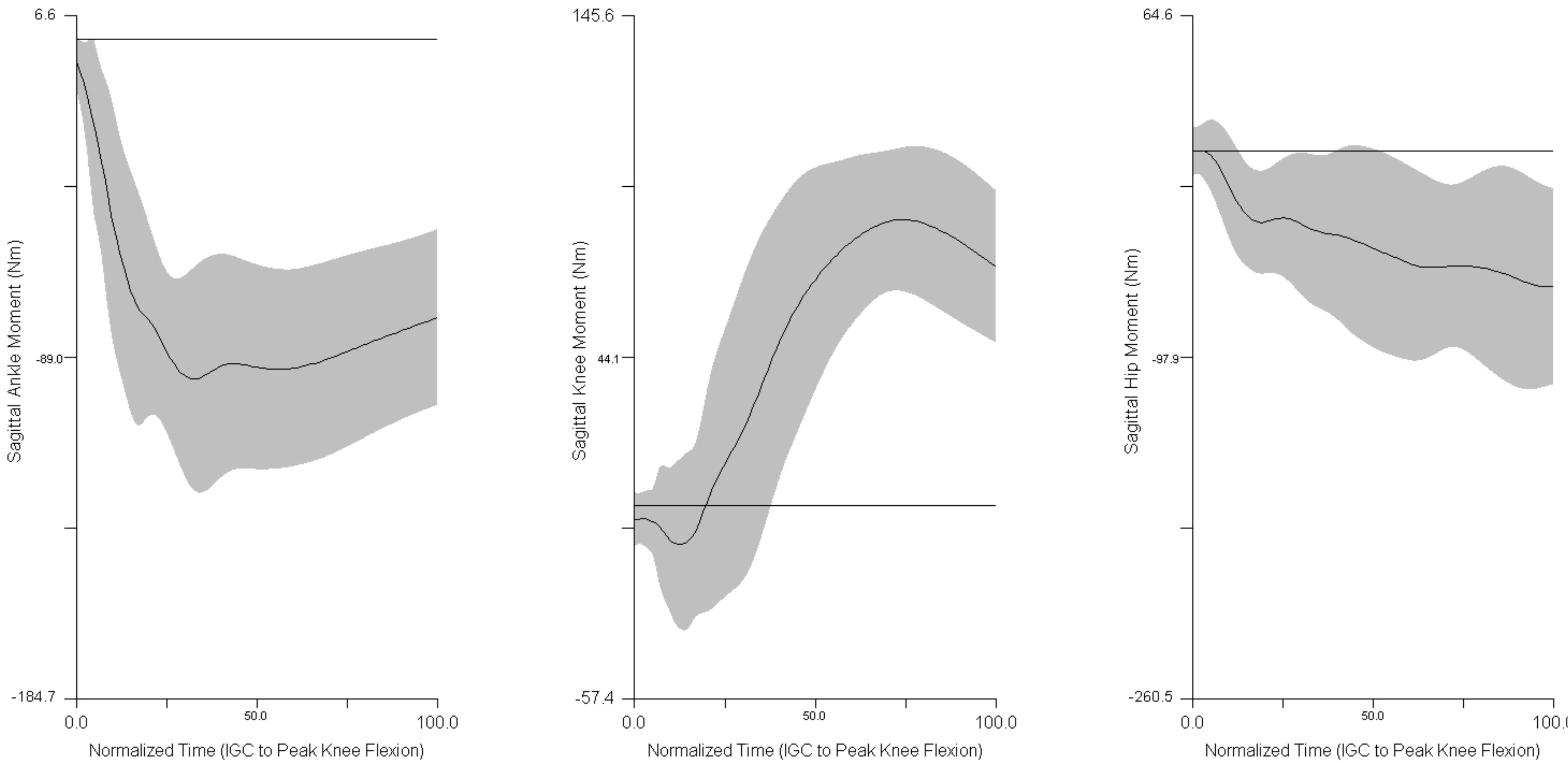


Figure K-7: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

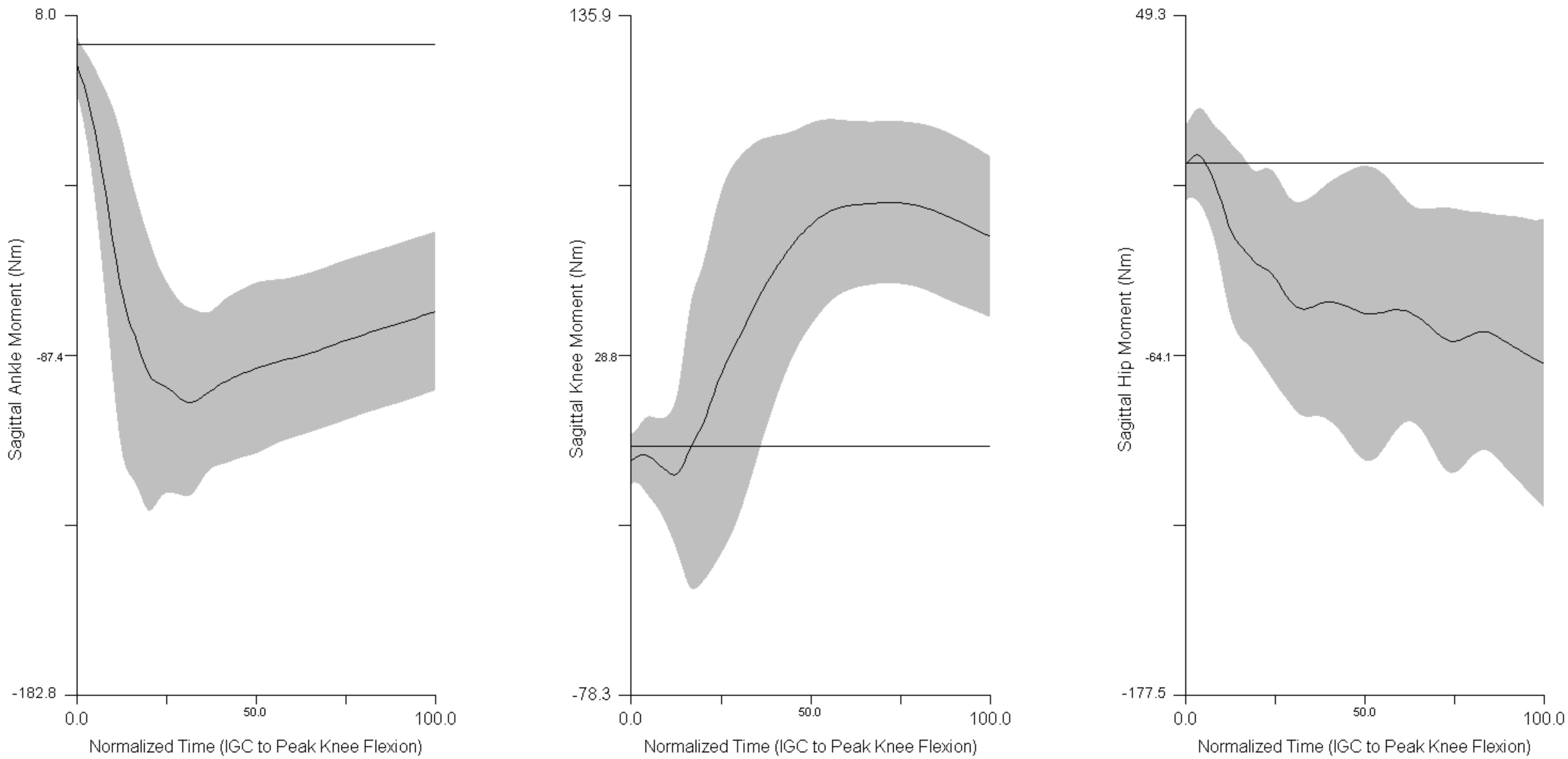


Figure K-8: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

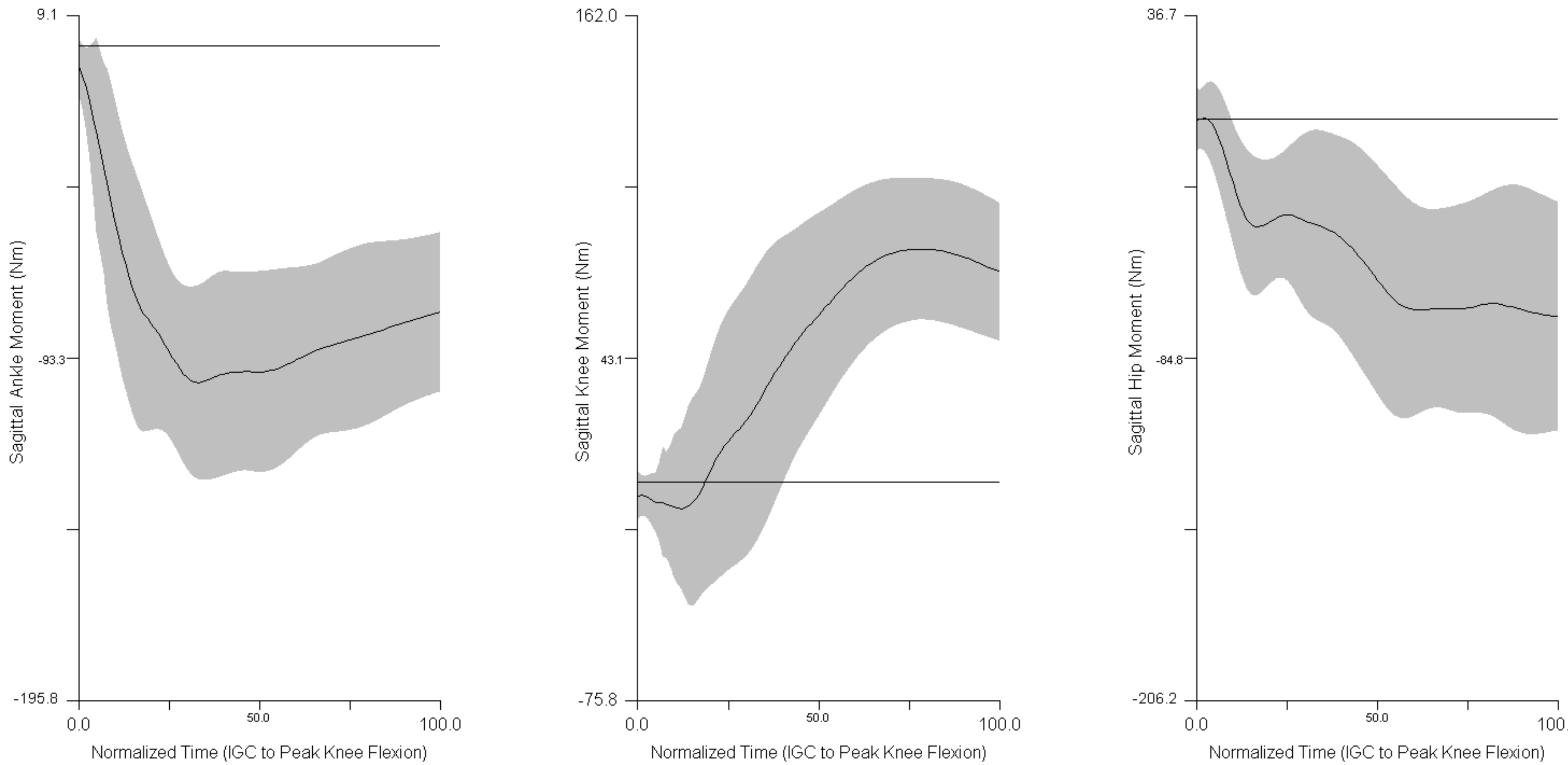


Figure K-9: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

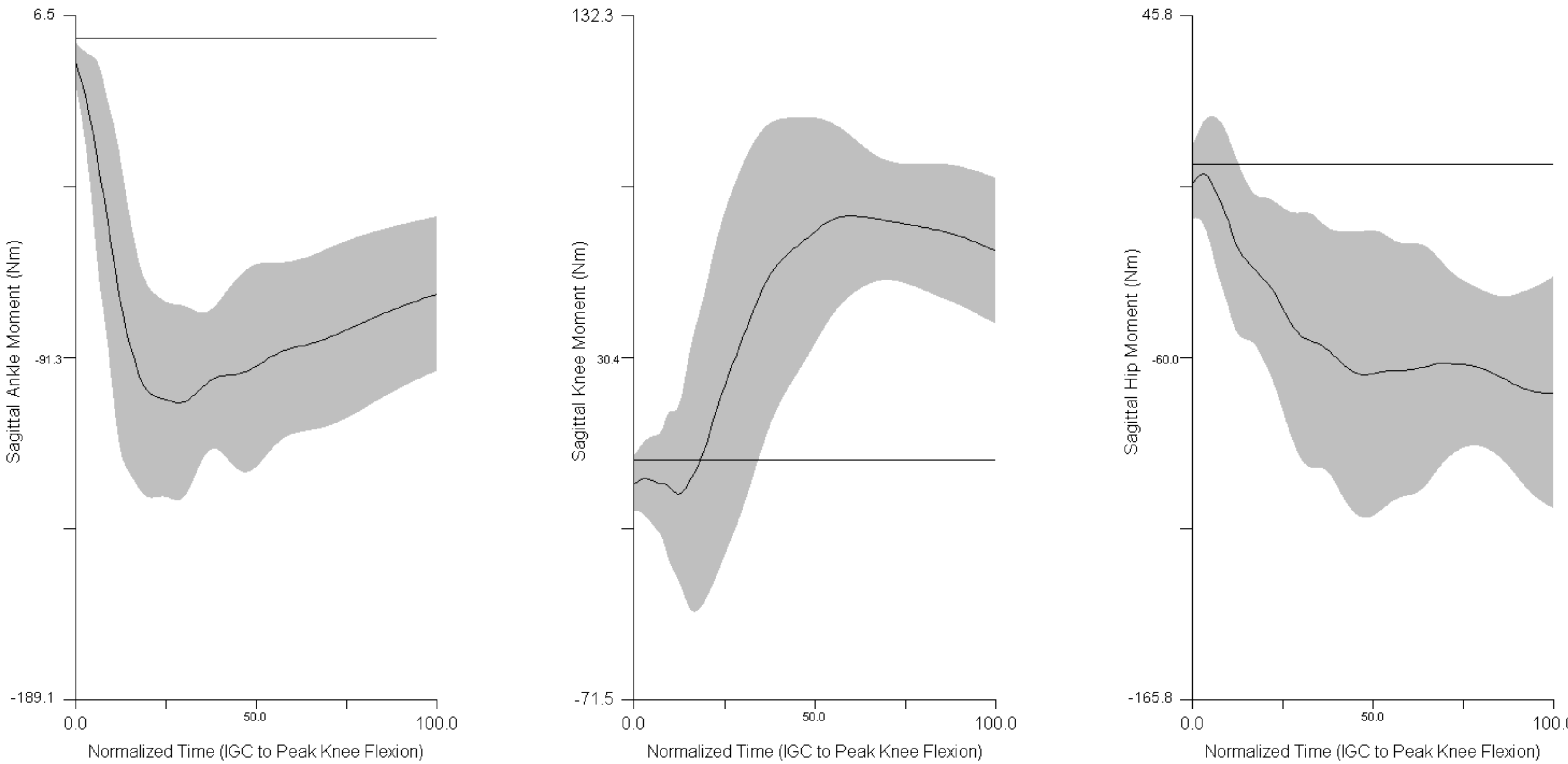


Figure K-10: Overall mean and standard deviation results for ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

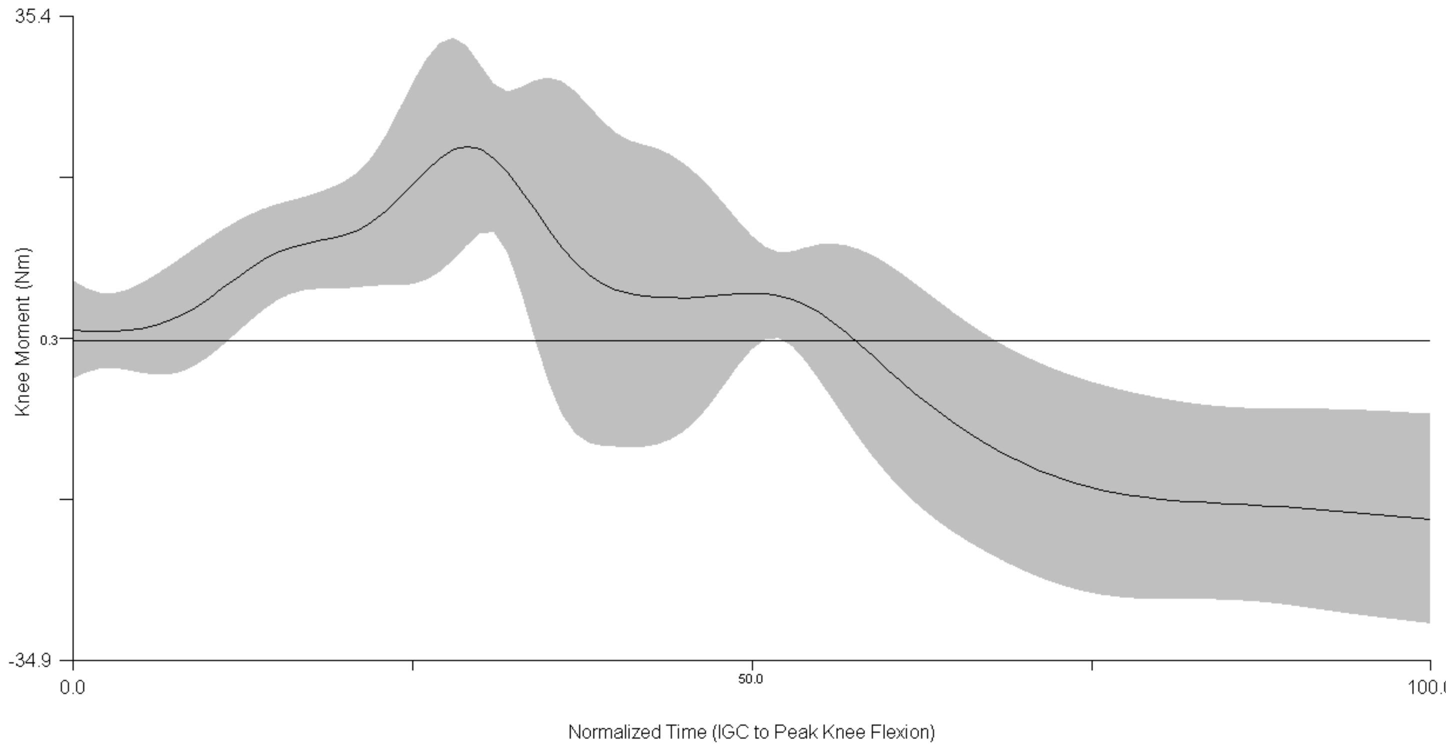


Figure K-11: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

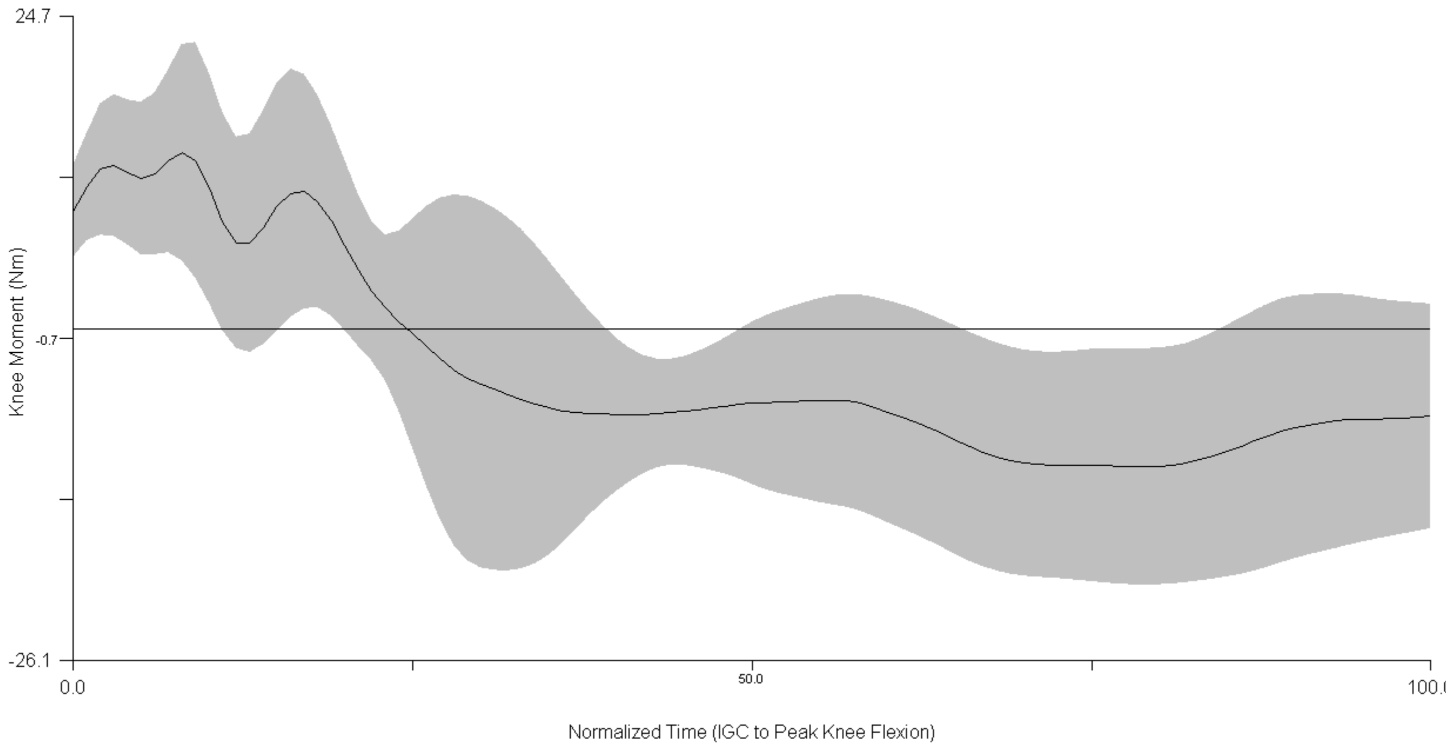


Figure K-12: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 1** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

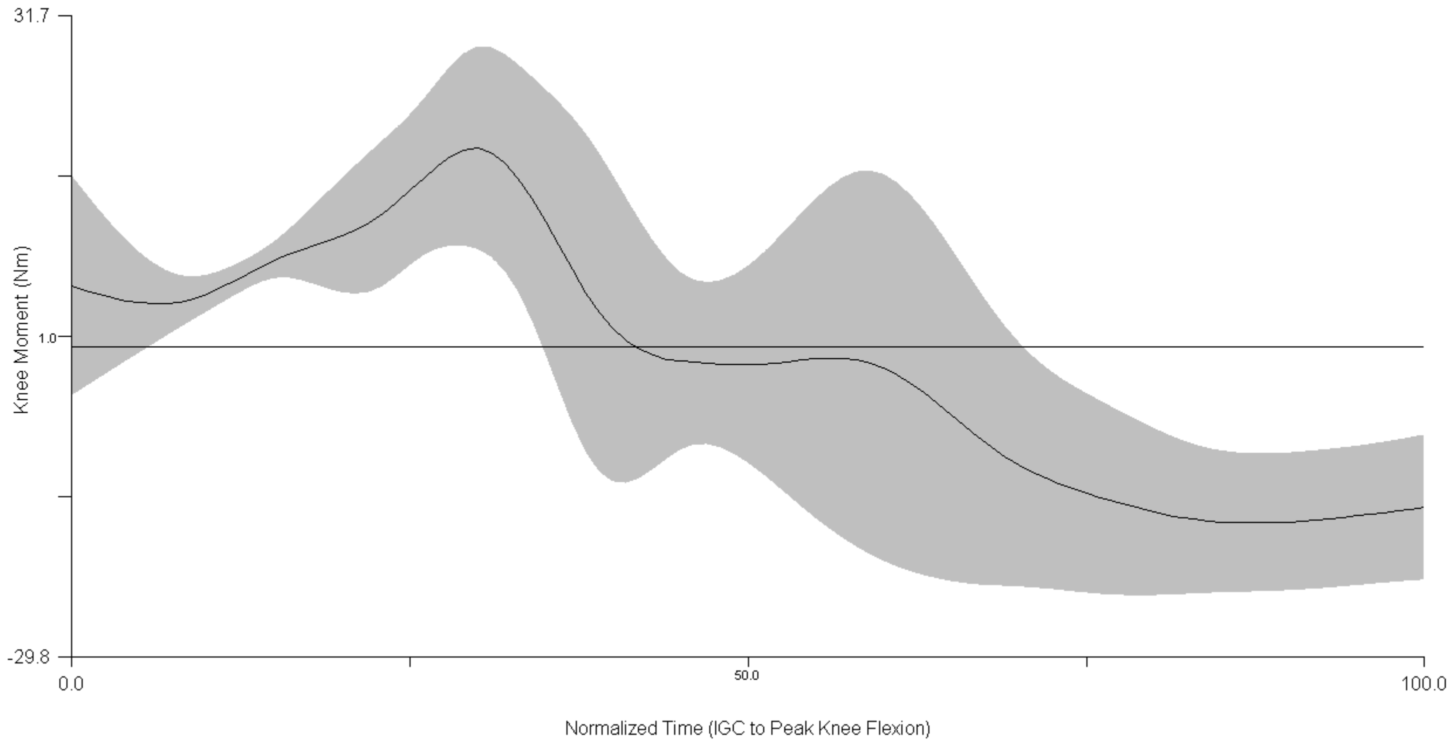


Figure K-13: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

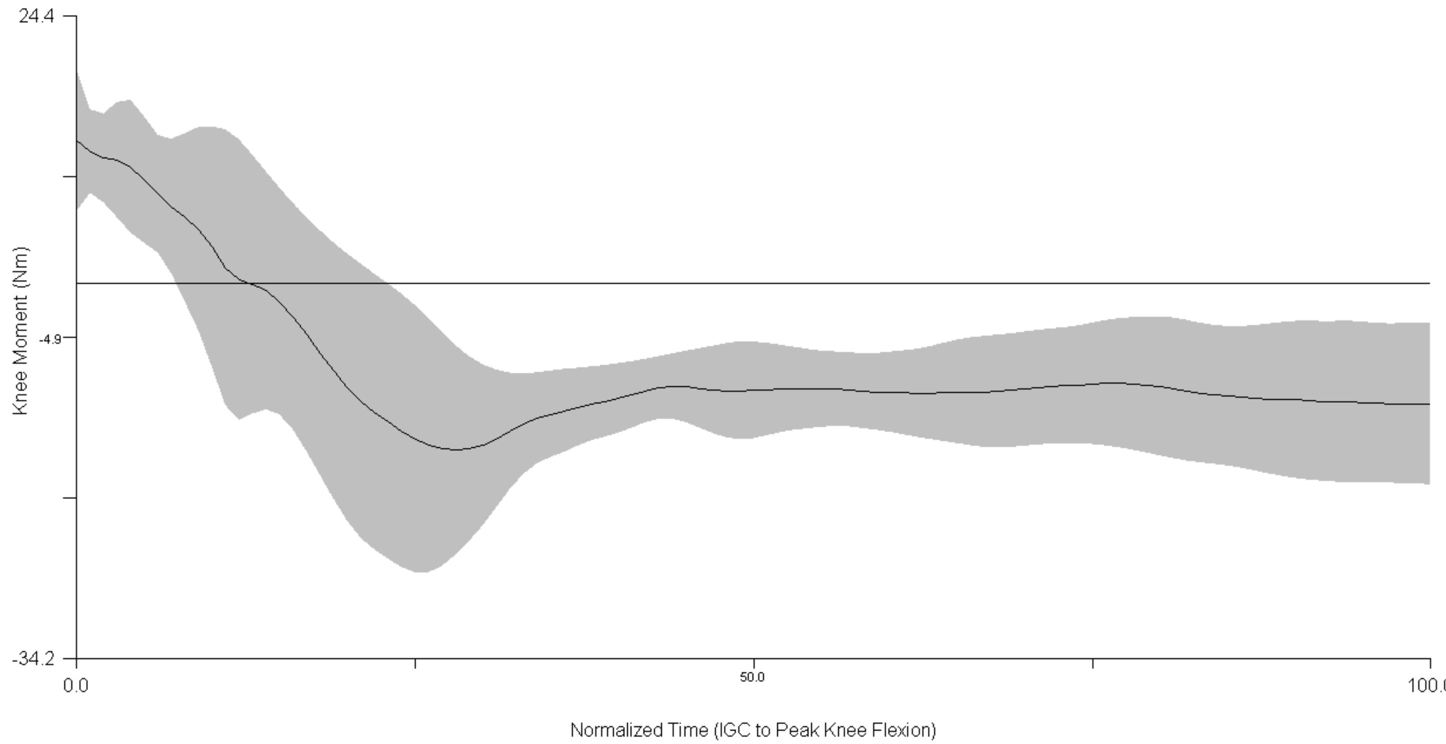


Figure K-14: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 2** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

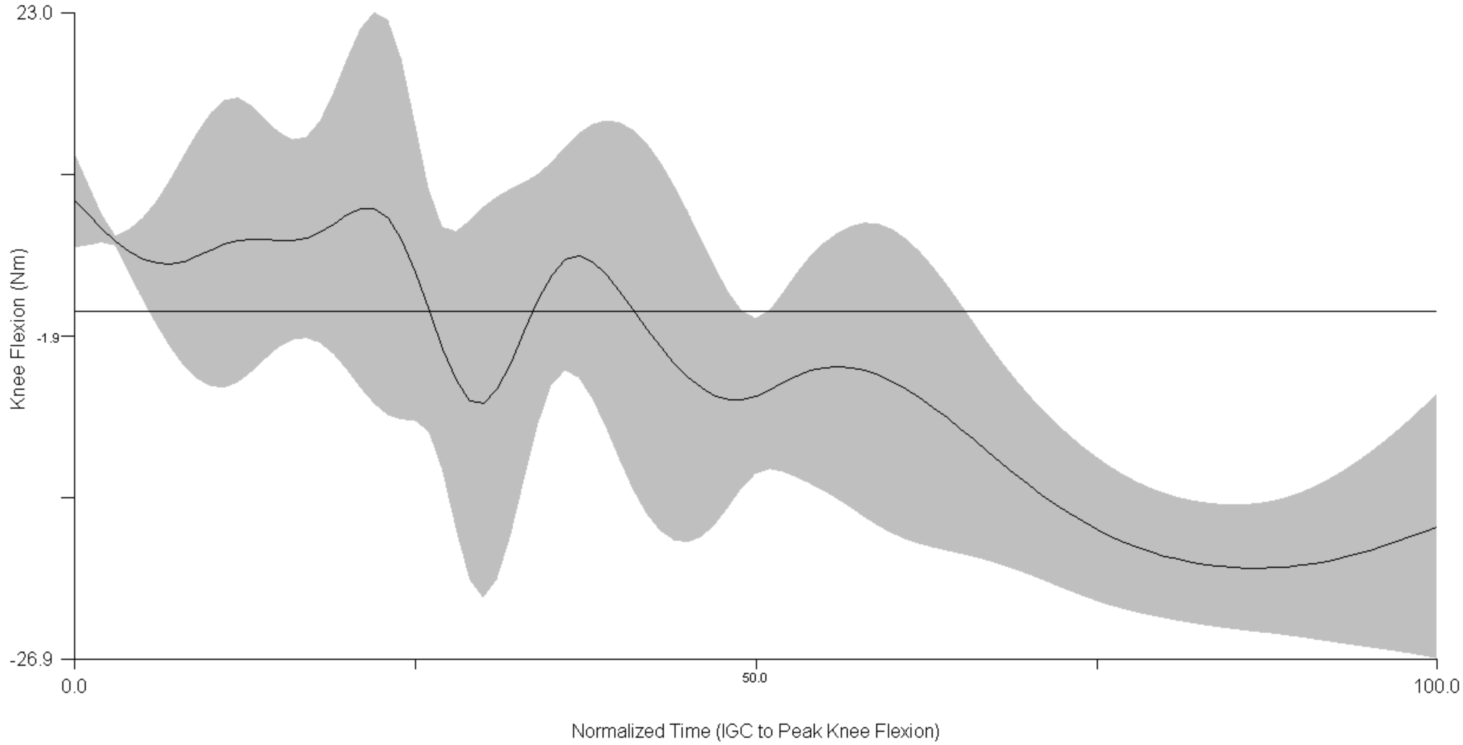


Figure K-15: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

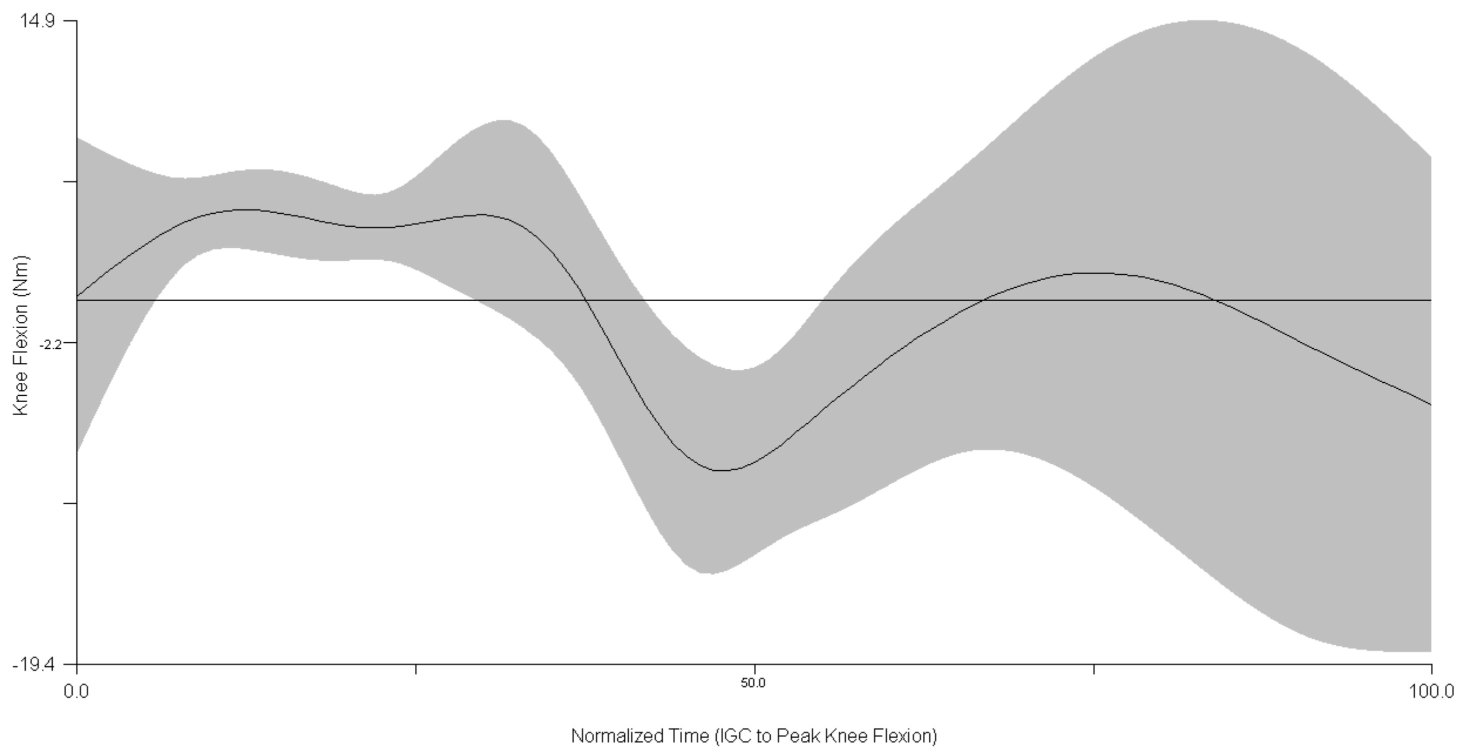


Figure K-16: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 3** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

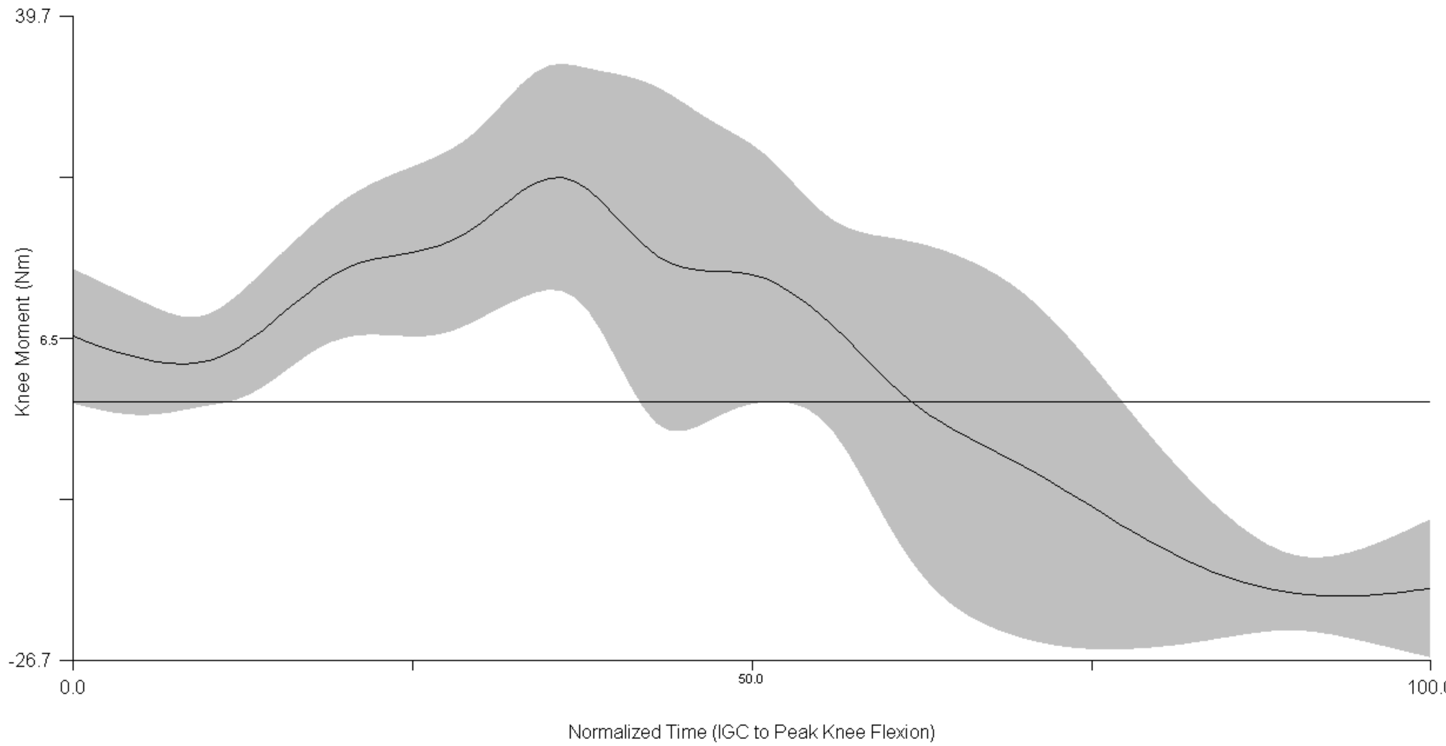


Figure K-17: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

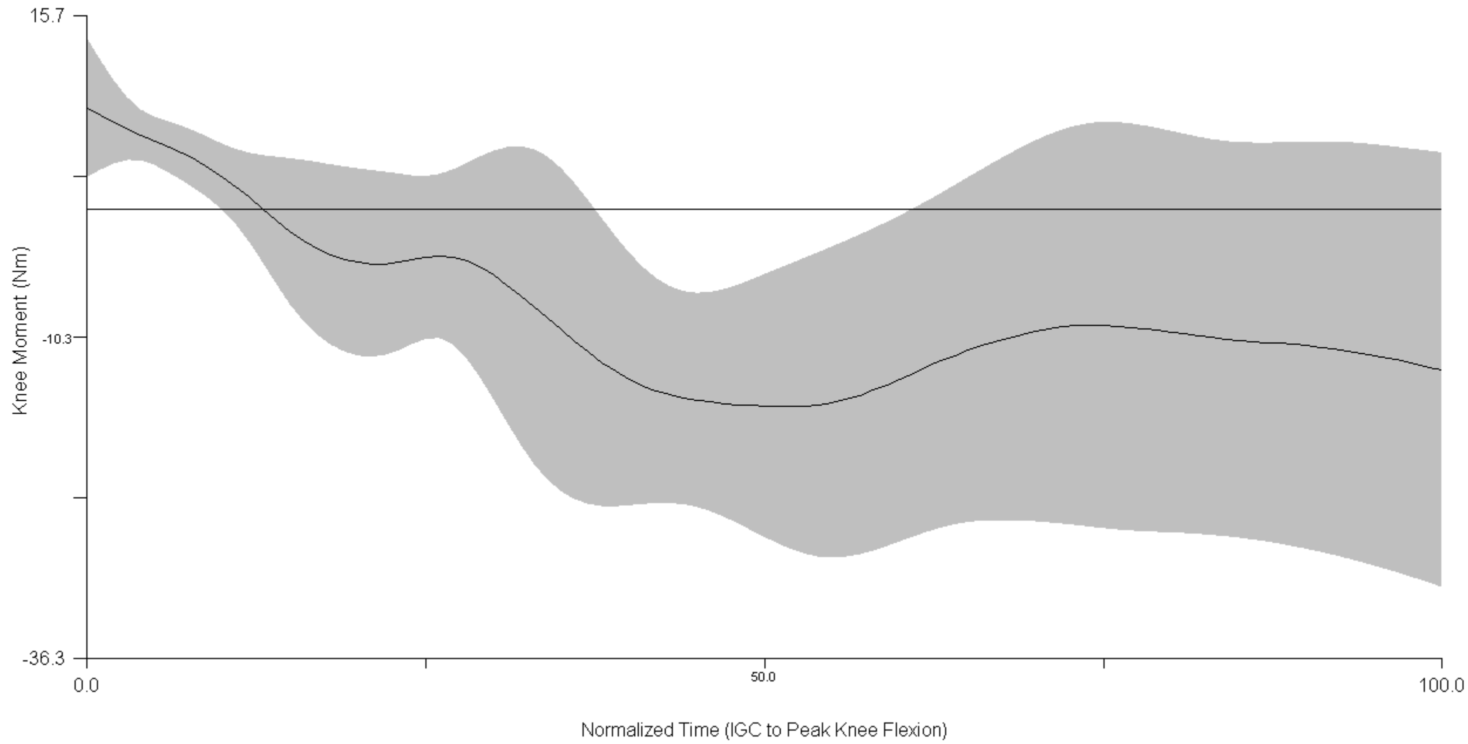


Figure K-18: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 4** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

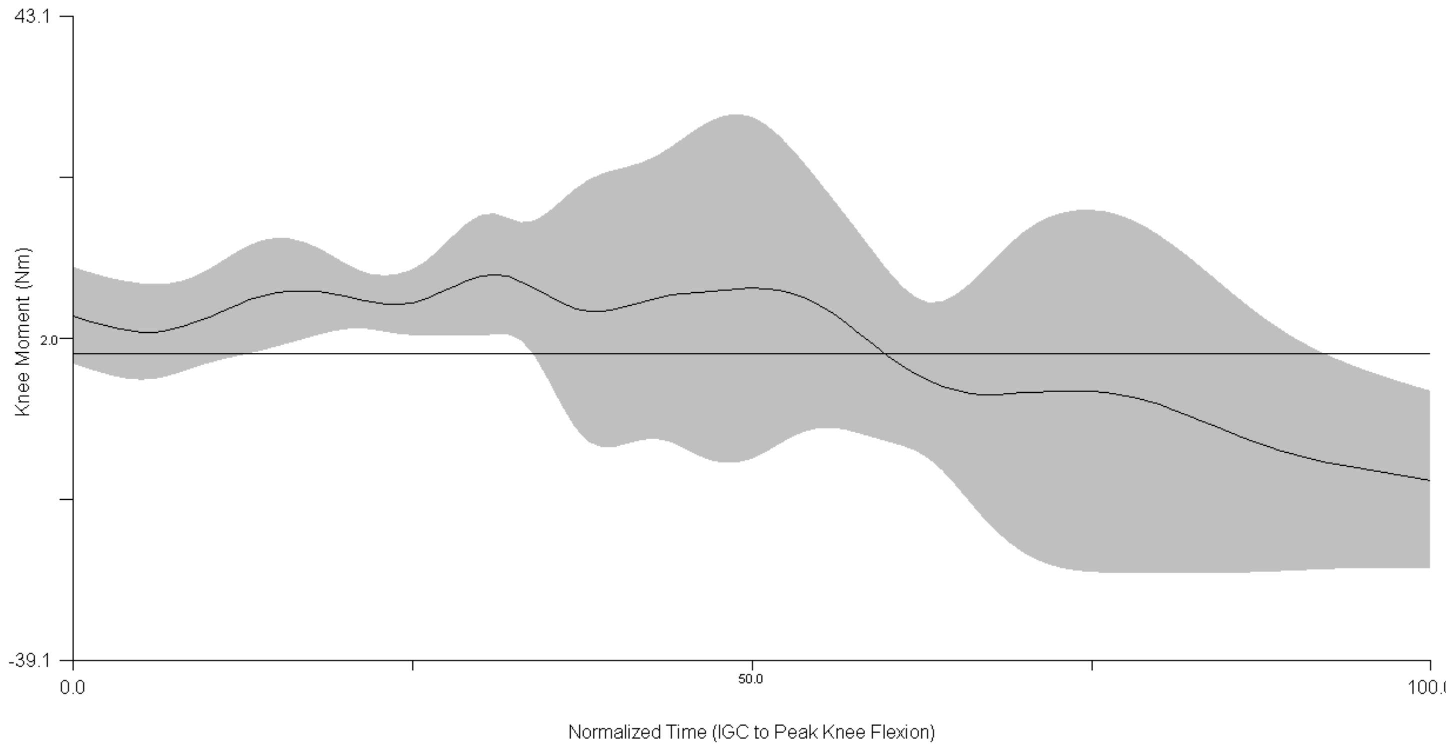


Figure K-19: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **unbraced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.

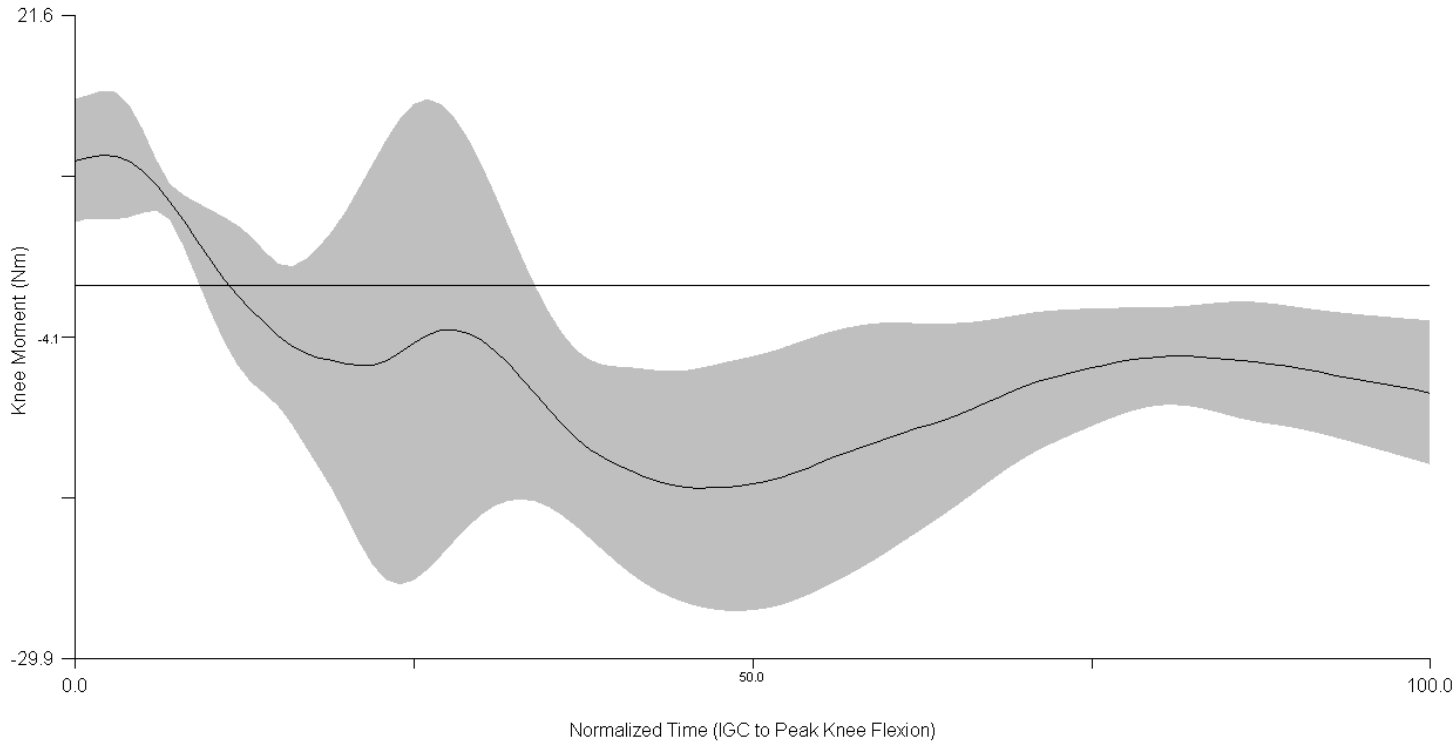


Figure K-20: Overall mean and standard deviation results for *frontal plane* ankle (left), knee (centre), and hip (right) moments for **braced** trials for **Time 5** across all twelve participants. The x-axis is normalized time from initial ground contact to peak sagittal knee flexion.