Influence of Chronic Ankle Instability on Human Movement: A Three Dimensional Kinematic and Electromyographic Analysis

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Abstract

Context: Lateral ankle sprains are one of the most common musculoskeletal injuries in the general and sporting population and as such present high-cost implications and time lost to sport and employment. Following an initial lateral ankle sprain, a high percentage of people develop chronic ankle instability with symptoms such as reduced range of motion, strength and proprioceptive deficits, episodes of giving way and instances of re-injury. Research investigating full body with multi-segmental foot kinematics and electromyography is limited thus impacting the development of successful rehabilitation and injury prevention strategies. **Aim:** The purpose of this research was to perform exploratory kinematic and surface electromyographic (sEMG) data analysis of the trunk, hip, knee, forefoot-tibia, forefoot-hindfoot and hindfoot-tibia between individuals with chronic ankle instability and healthy controls during walking, landing and cutting, three movements commonly associated with lateral ankle sprains **Participants:** Eighteen (14 males, 4 females) healthy controls (age 22.4 ± 3.6 years, height 177.8 ± 7.6 cm, mass 70.4 ± 11.9 kg) and 18 (13 males, 5 females) participants with chronic ankle instability (age 22.0 ± 2.7 years, height 176.8 ± 7.9 cm, mass 74.1 ± 9.6 kg). Participants' data were split into the healthy control and chronic ankle instability groups based on the results of the Identification of Functional Ankle Instability questionnaire. **Methods:** Participants were tested during walking (Chapter 6.0), single leg landing (Chapter 7.0) and cutting (Chapter 8.0). Three-dimensional kinematics were collected using the combined Helen Hayes and Oxford Foot Model and sEMG recorded for the peroneus longus, tibialis anterior and gluteus medius. Statistical parametric mapping, discrete variable analysis and regression analysis were subsequently performed. **Results:** Significantly modified kinematics were observed in each of the movements performed in the chronic ankle instability group.

Decreased forefoot-tibia internal rotation angular displacement was found to occur prior to initial contact in all three of the observed movements when comparing the affected limb to the healthy matched control prior to initial contact. Significantly modified electromyography was observed in the chronic ankle instability group during the cutting manoeuvre but not during the walking and landing manoeuvre. **Conclusions:** Key differences have been observed between groups specific to movements but also across movements. These differences are identified in not just foot and ankle kinematics but also higher up the kinetic chain in the knee, hip and trunk. Decreased forefoot-tibia internal rotation may be a variable of interest for future research due to its presence in each of the observed movements. Differences are also highlighted in the contralateral limb of the chronic ankle instability. These findings may therefore be used in the development of injury prevention and rehabilitation programmes and in the development of screening strategies. This could help to aid in the reduction in incidence of chronic ankle instability and improve the quality of life for those with chronic ankle instability.

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Abbreviations

1D	One-Dimensional
2D	Two-Dimensional
3D	Three-Dimensional
AII	Ankle Instability Instrument
AJFAT	Ankle Joint Functional Assessment Tool
ATFL	Anterior Talofibular Ligament
CAI	Chronic Ankle Instability
CAIS	Chronic Ankle Instability Scale
CAIT	Cumberland Ankle Instability Tool
CFL	Calcaneofibular Ligament
EMG	Electromyography
FAAM	Foot and Ankle Ability Measure
FAI	Functional Ankle Instability
FAIQ	Foot and Ankle Instability Questionnaire
FAOS	Foot and Ankle Outcome Score
FFHFA	Forefoot-Hindfoot Angle
FFTBA	Forefoot-Tibia Angle
FS	Foot Strike
FVA	Foot Velocity Algorithm

GRF	Ground Reaction Force
HFMM	Heidelberg Foot Measurement Method
HFTBA	Hindfoot-Tibia Angle
HS	Heel Strike
IAC	International Ankle Consortium
IC	Initial Contact
IdFAI	Identification of Functional Ankle Instability
MAI	Mechanical Ankle Instability
MFM	Milwaukee Foot Model
MVIC	Maximal Voluntary Isometric Contractions
OFM	Oxford Foot Model
PTFL	Posterior Talofibular Ligament
RMS	Root Mean Square
SD	Standard Deviation
sEMG	Surface Electromyography
SENIAM	Surface Electromyography for the Non-Invasive Assessment of Muscles
SPM	Statistical Parametric Mapping

Chapter 1.0 Introduction

Ankle sprains are one of the most common musculoskeletal injuries in a sporting population and the general population (Fong, Hong, Chan, Yung, & Chan, 2007; Gribble et al., 2016). Lateral ankle sprains are thought to occur with inversion or inversion and plantarflexion (Seah & Mani-Babu, 2011; Waterman, Belmont, Cameron, Deberardino, & Owens, 2010) and it has been reported that ankle sprains account for 60.9 admissions to Accident And Emergency Departments in the United Kingdom in every 10,000 persons each year (Bridgman et al., 2003). Following an acute ankle sprain up to 74% of individuals report chronic residual symptoms such as recurrent sprains, episodes of giving way and/or perceived instability, this is often referred to as chronic ankle instability (CAI)(Gribble et al., 2013). It is suggested that abnormal kinematic movement patterns adopted by those with CAI may increase repetitive cartilage damage to the medial ankle, thus long term links have been established between the development of osteoarthritis and history of CAI (Valderrabano, Hintermann, Horisberger, & Fung, 2006).

A number of rehabilitation and preventative strategies are adopted in an attempt to reduce the incidence of lateral ankle sprains from strengthening programmes (Calatayud et al., 2014; Wilkerson, Pinerola, & Caturano, 1997), balance and coordination training/unstable surface training (for example sand training or balance boards) - (Alghadir, Zafar, & Iqbal, 2015; McKeon & Mattacola, 2008; Ringhof, Leibold, Hellmann, & Stein, 2015), movement re-education (Caulfield & Garrett, 2004) and external supports in the form of rigid taping or kinesiology taping or bracing (McKeon & Mattacola, 2008; Shima, Maeda, & Hirohashi, 2005). However, despite these strategies, the rate of ankle sprains in individuals with ankle instability remains high. Increasing knowledge of biomechanical quantities is thought to be extremely important in the development of injury prevention strategies and protective equipment (Bahr & Krosshaug, 2005; Fong, Ha, Mok, Chan, & Chan, 2012). Understanding the differences in biomechanics may help to identify the risk for lateral ankle sprains in this susceptible population. Existing methods may be enhanced with increased knowledge of mechanisms and biomechanical quantities (Fong et al., 2012; Kristianslund, Bahr, & Krosshaug, 2011).

The kinetic chain principle states that a combination of successively arranged joints constitutes a complex unit and as such movement of one joint affects the movement of another (Karandikar & Vargas, 2011). Movement of the trunk (which accounts for 35.5% body mass) will also have an impact on the motion of the hip and therefore knee and ankle (Kulas, Zalewski, Hortobagyi, & DeVita, 2008). Research has suggested that individuals utilise either an ankle or hip strategy (Horak & Nashner, 1986; Kuo & Zajac, 1993; Runge, Shupert, Horak, & Zajac, 1999). The ankle strategy involves moving the body about the ankle joint whilst the hip strategy utilises movement of the hip in order to maintain the centre of gravity within the base of support (Horak & Nashner, 1986). Individuals with CAI have been reported to adopt a 'top-down' or hip strategy to maintain balance (Hubbard, Kramer, Denegar, & Hertel, 2007; Tropp, Odenrick, & Gillquist, 1985). To produce the most effective intervention strategy for individuals with ankle instability it is essential to have knowledge of the full kinetic chain so it is clear the benefit an intervention will have on other surrounding joints. Research to date has already documented the implications of proximal adaptations that exist with simple taping and bracing interventions (Cordova, Takahashi, Kress, Brucker, & Finch, 2010; DiStefano, Padua, Brown, & Guskiewicz, 2008; Santos, McIntire, Foecking, & Liu, 2004; Stoffel et al., 2010). No current research investigates full body kinematics and muscle activation patterns during dynamic movements prone to injury.

This thesis will adopt an exploratory study design to further understand full body kinematics across full-time series and at discrete time points during human movement to better understand the biomechanics adopted by those individuals with CAI that are prone to recurrent sprains and episodes of giving way. It will also adopt an electromyographic analysis of key muscles to identify differences in patterns of muscle activation. A deeper understanding of muscle activation and kinematics may inform more appropriate preventative measures to be adopted in future research.

1.1 Contributions to the Literature

Research from within this thesis has been presented in the following formats:

Peer-reviewed publications

Northeast, L., Gautrey, C. N., Bottoms, L., Hughes, G., Mitchell, A. C., & Greenhalgh, A. (2018). Full gait cycle analysis of lower limb and trunk kinematics and muscle activations during walking in participants with and without ankle instability. *Gait & Posture, 64*(1), 114-118.

Conference Communications

Northeast, L., Gautrey, C., Mitchell, A., Bottoms, L. & Greenhalgh, A. (2018). A comparison of lower limb kinematics and electromyography during walking between athletes with chronic ankle instability and healthy controls. World Congress of Biomechanics, Dublin, Ireland, 8-12 July 2018.

Northeast, L., Gautrey, C., Mitchell, A., Bottoms, L. & Greenhalgh, A. (2018). A comparison of lower limb angular displacements, velocities and accelerations during walking between athletes with chronic ankle instability and healthy controls. World Congress of Biomechanics, Dublin, Ireland, 8-12 July 2018.

Several other papers are also in the process of being submitted for peer-reviewed publication in the near future.

Chapter 2.0 Review of Literature

2.1 Anatomy of the Ankle

The human foot is a very complex structure consisting of 26 bones and 33 joints (Tomassoni, Traini, & Amenta, 2014). Within research and clinical settings, the foot is normally divided into three parts; the forefoot (phalanges and metatarsals), hindfoot (calcaneus) and midfoot (cuneiforms, cuboid and navicular) which forms the arch of the foot (Tomassoni et al., 2014).

The ankle consists of three main articulations - the talocrural joint, the subtalar joint and the distal tibiofibular syndesmosis (Hertel, 2002). The talocrural joint allows for plantarflexion and dorsiflexion, around its articulations (the dome of the talus, the tibial plafond and the medial and lateral malleolus) (Fong et al., 2009a). The subtalar joint articulations are the plantar surface of the talus and the calcaneus allowing inversion and eversion to take place (Dubin, Comeau, McClelland, Dubin, & Ferrel, 2011). The distal tibiofibular joint articulations are the distal tibia and fibula which allow for slight accessory gliding (Hertel, 2002). The fibula glides superiorly and rotates laterally with dorsiflexion in order to allow the anterior aspect of the talus to move into the mortise, with plantarflexion, the opposite must occur with the fibula gliding inferiorly and internally rotating (Loudon & Bell, 1996).

The ligaments surrounding the ankle include the lateral collateral ligaments (Figure 2.1) (anterior talofibular ligament (ATFL), calcaneofibular ligament (CFL) and the posterior talofibular ligament (PTFL)), the medial collateral ligaments also referred to as the deltoid ligament (Figure 2.2) (tibionavicular, tibiocalcaneal and the tibiotalar ligaments) and the syndesmotic ligaments (anterior inferior tibiofibular ligament, interosseous ligament, posterior inferior tibiofibular ligament and the transverse ligament) (Dubin et al., 2011).

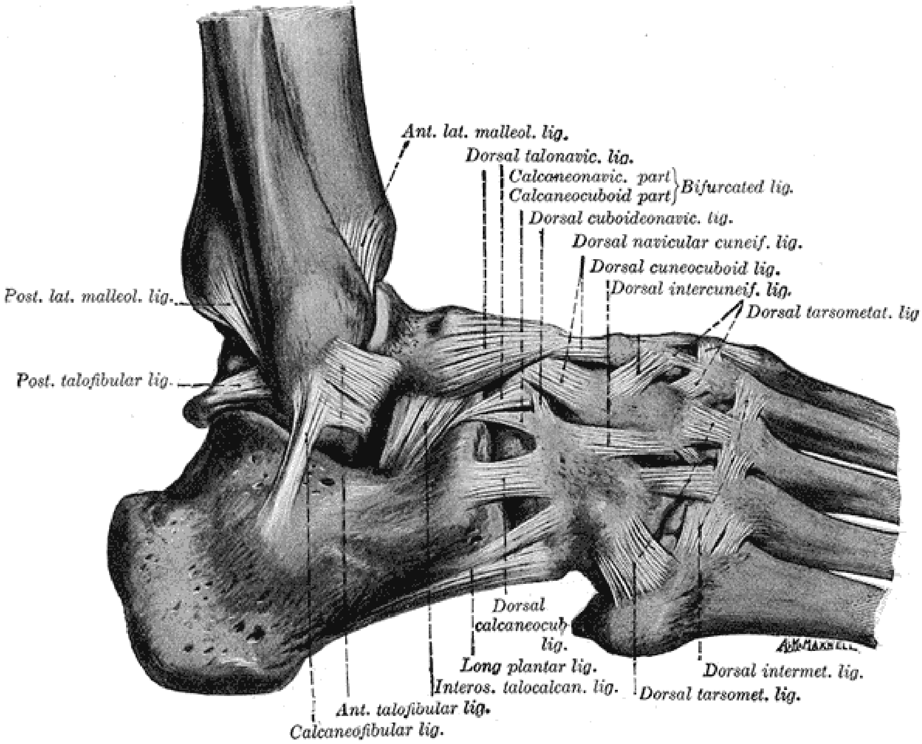


Figure 2.1 Lateral view of the ankle (Gray, Goss, & Alvarado, 1973).

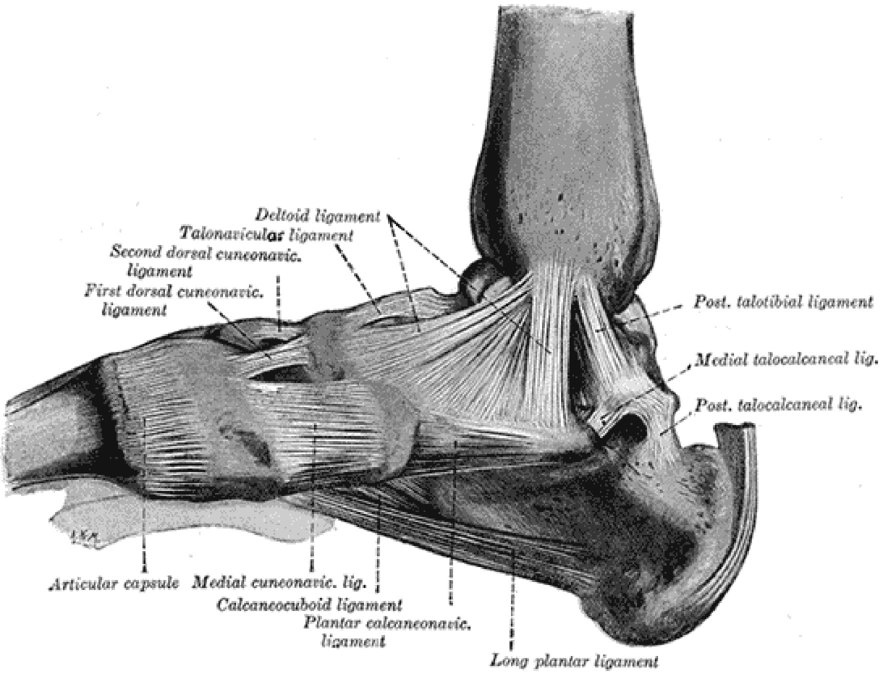


Figure 2.2 Medial view of the ankle (Gray et al., 1973)

The ATFL originates from the anterior margin of the fibular malleolus and inserts onto the lateral aspect of the neck of the talus (Dubin et al., 2011). The role of the ATFL is to limit anterior displacement of the talus and excessive plantarflexion of the ankle (Golanó et al., 2010). The CFL originates from the lateral malleolus and inserts onto a tubercle on the lateral calcaneal surface (Yıldız & Yalcın, 2013). The role of the CFL is to limit excessive inversion and motion in the frontal plane (Brown, Padua, Marshall, & Guskiewicz, 2008). The PTFL originates from the posterior aspect of the lateral malleolus and inserts into the posterolateral aspect of the talus and it helps to restrict excessive inversion and internal rotation of the loaded talocrural joint (Hertel, 2002). The PTFL is the strongest of the lateral ligaments and works to stabilise the talus when the ankle is in maximal dorsiflexion (Woodman, Berghorn, Underhill, & Wolanin, 2013).

The tibionavicular ligament is the most superficial part of the deltoid ligament originating at the anterior colliculus of the tibia and inserting to the dorsomedial surface of the navicular (B. R. Williams, Ellis, Yu, & Deland, 2010). The tibiocalcaneal ligament again originates from the anterior colliculus and attaches to the sustentaculum tali (Beals, Crim, & Nickisch, 2010). The deep portion of the deltoid ligament is made up of the anterior and posterior tibiotalar ligaments (Chhabra, Subhawong, & Carrino, 2010). The deep portion originates from the intercollicular groove and inserts on the medial talus, the posterior ligament is the thickest portion of the medial ligament complex (B. R. Williams et al., 2010). The deltoid ligaments work together to resist excessive external rotation and eversion of the ankle joint (Dubin et al., 2011).

The ankle musculature is extremely important in the protection of lateral ankle sprains particularly the peroneus longus and brevis muscles (Delahunt, Monaghan, & Caulfield, 2006a). The anterior compartment consisting of the tibialis anterior, extensor digitorum longus and brevis and peroneus tertius also contribute to dynamic stability by slowing plantarflexion and inversion (Hertel, 2002).

2.2 Mechanisms of Lateral Ankle Sprains

The commonly accepted mechanism for lateral ankle sprains is inversion and plantarflexion whilst medial ankle sprains involve eversion and dorsiflexion (Seah & Mani-Babu, 2011; Waterman et al., 2010). In contrast, some research suggests that internal rotation along with inversion and not the previously suspected plantarflexion may be the mechanism for lateral ankle sprains (Mok et al., 2011).

Lateral ankle sprains commonly occur when the centre of gravity is shifted laterally over the lateral border of the weight bearing foot causing high-velocity inversion to occur (Dubin et al., 2011). They generally occur in activities such as cutting and jump landing, in sports involving jumping the player may land on another player's foot or catch the lateral edge of the foot causing excessive ankle inversion to occur (Robbins & Waked, 1998). It has also been suggested that inversion sprains may occur due to poor positioning of the foot before and at foot contact due to a loss of proprioceptive input from mechanoreceptors (Willems, Witvrouw, Verstuyft, Vaes, & De Clercq, 2002).

The ATFL is the weakest of the ankle ligaments with an ultimate load of 138.9 N ± 23.5 N in comparison to 345.7 N ± 55.2 N displayed for the CFL (Attarian, Mccrackin, Devit, Mcelhaney, & Garrett, 1985). The ATFL is the most commonly injured ligament in the ankle complex with research showing damage to have occurred in 82.8% of ankle injuries (Fallat, Grimm, & Saracco, 1998). In more severe ankle sprains, damage to the CFL may also be present (Dubin et al., 2011). This is found to be the case in 66.9% of ankle injuries (Fallat et al., 1998). Anterior talofibular ligament sprains occur with plantarflexion and inversion whilst the CFL is sprained with dorsiflexion and inversion or just inversion (Konradsen & Voigt, 2002). Previous links have been made with ruptured ATFL and CFL suggesting the mechanism starts with a plantarflexion and inversion position but when being forced into further inversion, rupture of the ATFL occurs, if this continues the foot is forced into dorsiflexion with further inversion before eventual rupture of the CFL occurs (Konradsen & Voigt, 2002).

Other structures can be injured during ankle sprains such as the lateral joint capsule, proprioceptive nerve endings and the peroneal tendons (Dubin et al., 2011). To effectively reduce the risk of ankle sprains a full understanding of the injury mechanism and kinematic and kinetic injury mechanisms is crucial.

2.3 Epidemiology of Ankle Sprains

A study surveying UK Accident and Emergency Departments in Dudley, Sandwell, Walsall and Wolverhampton hospitals found ankle sprains to account for 60.9 in every 10,000 persons (Bridgman et al., 2003). If this is applied to the populations of England and Wales it is estimated this will account for 302,000 sprains annually. Injuries to the ankle are reported to account for 20-40% of all athletic injuries (Dubin et al., 2011). Further to this a study investigating attendance at Accident and Emergency Departments in the UK assessed 1,715 sport injury cases between 1st January 2005 and 31st December 2005 finding 240 ankle injuries of which 81.3% were ligamentous sprains (Fong, Man, Yung, Cheung, & Chan, 2008). A study investigating ankle sprains in an Emergency Department in the south of England over a seven month period found the causes of ankle sprains to be tripping (29%), non-specific injury (26.4%), playing sport (26%), walking (12.2%) and other accidental causes (6%) (Al Bimani et al., 2018). It has been stated that up to one-sixth of the time lost from sporting activities is caused by ankle sprains (Rein, Fabian, Zwipp, Heineck, & Weindel, 2010). This injury rate could be significantly underestimated as it has been reported that 55% of people suffering from lateral ankle sprains do not seek care from health care professionals (Wikstrom, Tillman, & Borsa, 2005). A high-cost implication has been observed from those that do seek help with an American study showing direct medical costs incurred following ankle sprains accounted for a total of $70 billion and indirect costs amounted to $1.1 billion in 2003 alone (McGuine & Keene, 2006). In the UK it has been calculated that if all severe ankle sprains were treated with a boot this would equate to an expenditure of £3 million, for a brace £1.5 million and for just a tubigrip £0.1 million per year (Bridgman et al., 2003). Ankle sprains are evidently a widespread issue therefore even a small decrease in occurrence will result in a significant decrease in time lost from sport and everyday activities, as well as a substantial economic saving.

Medial ankle sprains and syndesmotic or high ankle sprains have been found to account for 11.8% of all ankle sprains in a young, athletic population (Waterman et al., 2011). In comparison, lateral ankle sprains have been found to account for 77% of ankle sprains and in 73% of cases damage to the ATFL is involved (Fong et al., 2007). Ligament sprains are graded for severity with grade 1 being a mild stretch of the ligament without accompanying joint instability, grade 2 a partial rupture with mild instability whilst a grade 3 is a complete rupture with joint instability (Seah & Mani-Babu, 2011).

Fong et al. (2008) investigated sport related ankle sprains in the Accident and Emergency Department of the Prince of Wales Hospital in Hong Kong. They observed highest incidence rates in basketball and football - sports that require sudden cutting manoeuvres, jumping, landing and sudden stops. The next highest incidence rate was in hiking, potentially due to unstable surfaces. During activities like basketball and football, the athlete is at risk of catching the lateral edge of the foot or landing on another player's foot and causing the ankle to roll into combined inversion and plantarflexion (Knight & Weimar, 2011b; McGuine & Keene, 2006). Bahr and Krosshaug (2005) used a multifactorial approach to analysing ankle sprains. This included intrinsic and extrinsic risk factors, details of the event leading to the injury and a biomechanical description of the whole body and joint at the point of injury. They found ankle sprains in volleyball to mainly occur at the net when a player is blocking or attacking, they then landed on another player's foot. In football however these injuries most commonly occurred in late tackles due to a medial contact from an opponent to the leg before or at foot strike (FS).

A systematic review performed by Fong et al. (2007) reported injury trends from 227 studies including 70 sports from 38 countries. The results obtained showed the ankle to be the most commonly injured site in 24 of the 70 sports. They analysed the incidence of ankle sprains per 1000 person-hours. Rugby was found to have the highest incidence (4.20) followed by football (2.52) and volleyball (1.99). In basketball, ankle sprains have been found to account for up to 45% of all injuries sustained (Anandacoomarasamy & Barnsley, 2005). Each of these sports is intermittent in nature and involves a high incidence of cutting and changes of direction as well as jump landings.

2.4 Risk Factors for Ankle Sprains

A number of intrinsic and extrinsic factors are thought to predispose an athlete to ankle sprains (Kerkhoffs et al., 2012). Extrinsic factors are thought to include competition level, shoe type and playing surface (Ramanathan et al., 2011). Intrinsic factors include height, body mass index, age, sex, weight, previous injury, flexibility, muscle strength, proprioception, reaction time, anatomical alignment and postural stability (Waterman et al., 2010).

Intercollegiate athletes have been found to be twice as likely to sustain an ankle sprain than intramural athletes potentially due to increased competition intensity, increased aggression of contact, increased match exposure, limited rest periods and increased risk-taking (Waterman et al., 2011). More injuries have been found to occur in matches when compared to training sessions possibly due to the increased speed of play and increased contact (Wong & Hong, 2005).

There is speculation as to whether the type of shoe and shoe height affects the risk of ankle sprain, however, no significant effects have been identified: it has therefore been suggested that the age of the footwear has more of an impact (Verhagen & Bay, 2010). Increasing shoe sole thickness has been found to increase the protective eversion response of the peroneus longus, in order to counter the increasing moment of the subtalar joint during sudden inversion (Ramanathan et al., 2011) which signifies an increased risk with an increased sole thickness. When a grass pitch is warmer, harder and drier there is increased risk of ankle sprain due to an increased shoe-surface friction and an increased rate of increase in ground reaction force (Orchard & Powell, 2003). Shoe-surface friction has also been hypothesised to be higher on synthetic materials than on natural materials, therefore, increasing the risk of ankle sprains. An example of this is artificial turf in comparison to grass. Ekstrand, Timpka, & Hägglund (2006) compared injury rates over a season for 290 football players from 10 elite European clubs competing on a third-generation artificial turf to 202 players from the Swedish premier league. They reported an increased risk of ankle sprains on artificial turf, however, the overall incidence rate was very low and that further study would be required to draw conclusions from this. Another football based study by Steffen, Andersen and Bahr (2007) analysed injuries for 2,020 female players over a season their results showed a trend towards an increase in ankle sprains on artificial turf when compared to grass but found no significant differences.

Increased height of the athlete has also been suggested to be an intrinsic risk factor, it is hypothesised to cause an increased moment of inertia acting about the ankle joint therefore predisposing the athlete to an ankle sprain (Waterman et al., 2010). Moment of inertia is calculated using the formula: "moment of inertia = mass x perpendicular distance to the axis^2" therefore the taller the athlete the further the mass is distributed away from the axis leading to an increased moment of inertia (Waterman et al., 2010). Faude, Junge, Kindermann and Dvorak (2006) acquired baseline data, injury data and exposure times from 8 elite ladies football teams in the German national league. They found a risk factor of 9.64 for athletes greater than 1 SD above the mean height in contrast to 1.70 for athletes 1 SD below the mean height, however, their results show a high 95% confidence interval suggesting low precision of the odds ratio. Willems et al. (2005) collected the same data for 159 physical education students across 3 years and no relationship was found between athletes' height and incidence of ankle sprains. This could be due to a difference in experience level or the increased sample size.

It has been proposed that athletes with an increased body mass index are at greater risk of ankle sprains as they must generate greater forces to change momentum (momentum= mass x velocity) (Tyler, McHugh, Mirabella, Mullaney, & Nicholas, 2006). Greater forces are thought to be produced with heavier weights which must then be absorbed by the joints and soft tissue (Caine, Maffulli, & Caine, 2008). This is also thought to increase the risk of injury. McHugh, Tyler, Tetro, Mullaney, and Nicholas (2006) using 169 high school athletes (101 male and 68 female) found injury incidence to increase from 0.8 per 1000 exposures for male athletes to 3.0 per 1000 exposures with overweight males. This was however only seen with males and not females. The sample population consisted of 18 overweight and 19 at risk of overweight males in comparison to just 1 overweight and 6 at risk of overweight females, which may indicate why no correlation was seen in the female population.

Young athletes have been found to be prone to ankle sprains due to a decreased skill level and poorer risk evaluation (Cameron, Owens, & DeBerardino, 2010; Kofotolis & Kellis, 2007). Older athletes have however been found to have a worse recovery in terms of speed and quality following ankle sprains due to the age-related loss of strength and muscle mass (S. R. O'Connor, Bleakley, Tully, & McDonough, 2013).

Females have been found to have a 25% greater risk of ankle sprain than males in basketball (Hosea, Carey, & Harrer, 2000), possibly due to increased joint laxity, limb alignment, differences in total response time, ability to rapidly develop a lower extremity joint moment and decreased active muscle stiffness compromising joint stability. However Beynnon, Vacek, Murphy, Alosa, and Paller (2005) between 1999 and 2003 evaluated first time ankle sprains in high school and collegiate athletes in football, basketball, lacrosse and field hockey found that the risk was associated with the type of sport finding an increased risk only in basketball.

Athletes who have previously sustained an ankle sprain are twice as likely to suffer another than those who have not sprained their ankles previously (Steffen, Myklebust, Andersen, Holme, & Bahr, 2008). This has been proposed to be due to physiological and anatomical deficits such as reduced strength (Willems et al., 2002), decreased neuromuscular control (Gutierrez, Kaminski, & Douex, 2009) and increased joint laxity (Lentell et al., 1995). It is unclear whether an increased joint laxity results in late detection followed by a standard peroneal reaction, or if the movement is sensed at the normal time and a delayed peroneal reaction time occurs (Hoch & McKeon, 2014). Athletes who were found to have a decreased joint position sense at 15° inversion and 5°

from maximal inversion position were thought to be more at risk of ankle sprains (Willems et al., 2005). This was thought to be the case due to a lateral shift in the centre of pressure and poor foot positioning.

Limited dorsiflexion range of motion has been observed within individuals with ankle instability (Drewes, McKeon, Casey, & Hertel, 2009b). Decreased dorsiflexion has also been found to be a strong predictor of ankle sprain (Pope, Herbert, & Kirwan, 1998). A dorsiflexion range of 34° was found to have a five-fold greater risk of ankle sprain than an ankle of average flexibility (mean dorsiflexion 45 ± 4°) (Pope et al., 1998). It is proposed that this may be due to altered talocrural joint arthrokinematics where the talus is unable to glide posteriorly on the tibia and therefore increasing the risk of ankle sprains due to an inability to reach a stable closed-packed position (Drewes et al., 2009b). The knee to wall test is often used to test functional talocrural joint dorsiflexion range in a weight-bearing lunge position (Vicenzino, Branjerdporn, Teys, & Jordan, 2006). This is performed in standing with the heel in direct contact with the ground, with the knee passing in line with the second toe. Participants then lunge forward until their knee touches the wall. The foot is then gradually moved back until the point when the knee cannot touch the wall whilst still maintaining heel contact with the ground (Vicenzino et al., 2006).

Extrinsic risk factors associated with injury can often be easily modified. For example, ensuring athletes are competing at a suitable level or by changing the surface they are playing on. However, it is the intrinsic factors that are of particular concern as these cannot be modified (for example height, age and sex). With recurrent ankle sprains being reported in as high as 74% of individuals (Gribble et al., 2013) the costs of these in terms of time lost from sport and work and the costs to the local health care system are high.

This risk therefore needs to be addressed with more research so that suitable preventative strategies can be implemented.

2.5 Ankle Instability

Following an acute ankle sprain as high as 74% of individuals report residual symptoms such as recurrent sprains, episodes of giving way and/or perceived instability (Gribble et al., 2013). Chronic ankle instability is defined by Tanen et al. (2014) as 'a history of recurrent ankle sprains and the sensation of giving way'. The term CAI has been used in a number of studies though variation exists in the definition used. A position statement released by the International Ankle Consortium (IAC) outlined the definition that should be used in future research as 'an encompassing term used to classify a subject with mechanical and functional instability of the ankle joint' (Gribble et al., 2014).

Chronic ankle instability can be split into two categories: mechanical ankle instability (MAI) or functional ankle inability (FAI) (Wikstrom et al., 2005). These are not mutually exclusive and can occur individually and in combination (Brown et al., 2008; Wikstrom et al., 2005). Mechanical instability refers to an anatomical loss of mechanical restraint from tissues leading to an increase in joint laxity (Munn, Sullivan, & Schneiders, 2010; Wikstrom et al., 2005). This loss of restraint could be due to increased pathologic laxity, degenerative changes, synovial inflammation, impaired arthrokinematics and impingement (Hertel, 2002). It has been postulated, though neither theory has been confirmed, that mechanical instability may be caused by the increased motion of the talocrural joint or rotation of the talus within the ankle mortise leading to a rotary instability (Monaghan, Delahunt, & Caulfield, 2006).

Functional ankle instability was first proposed by Freeman, Dean and Hanham (1965). The term is used to describe a perception of weakness, pain, decreased functionality or giving way at the ankle joint (Hiller, Refshauge, Bundy, Herbert, & Kilbreath, 2006). The term "giving way" was clarified in the IAC position statement as 'the regular occurrence of uncontrolled and unpredictable episodes of inversion of the rear foot (usually experienced during initial contact (IC) during walking or running), which do not result in acute ankle sprains (Gribble et al., 2014). The exact cause of FAI is poorly understood (Caulfield & Garrett, 2004). It has been hypothesised that it may be due to failure of a dynamic restraint mechanism due to deficits in proprioceptive awareness, neuromuscular control, postural control and weakness of associated musculature (Hertel, 2002; Konradsen, Voigt, & Hojsgaard, 1997; Rein, Fabian, Zwipp, Rammelt, & Weindel, 2011; Rosen et al., 2013). The dynamic restraint mechanism is the speed at which support is provided to the joint complex by the contractile elements (Linford et al., 2006). Nyska et al. (2003) suggested that a leading cause of FAI is nerve injury either within or proximal to the ligament. Freeman (1965) proposed the deafferentiation theory, which stated that the afferent nerve fibres within the joint capsule and the ligaments of the foot and ankle stimulate reflexes which help to stabilise the foot during locomotion. These nerve fibres have lower tensile strength than collagen fibres and therefore, if the foot or ankle is sprained, partial deafferentiation of the injured joint occurs. Reflex stabilisation of the joint is then impaired which causes the joint to give way. Research following this theory has tended to focus most on proprioception, reflex and muscle response particularly of the evertors (peroneals) (Delahunt et al., 2006a; Eechaute, Vaes, Duquet, & Van Gheluwe, 2009; Monaghan et al., 2006).

Ankle instability has also been linked to the development of osteoarthritis (Valderrabano et al., 2006). Hip and knee osteoarthritis often affects older individuals, however, ankle arthritis in approximately 70-80% of cases is posttraumatic and as such is often prevalent in younger populations (Valderrabano et al., 2006). Research has found ligamentous lesions to be the cause of 13% of posttraumatic cases of osteoarthritis, with 85% being lateral ligament lesions (Valderrabano et al., 2006). One study used arthroscopic examination to investigate 148 patients with symptomatic CAI and reported cartilage lesions in 55% of cases and 62% of those to the medial aspect of the talus (Hintermann, Boss, & Schafer, 2002). It is suggested that abnormal kinematic movement patterns adopted by individuals with CAI may increase repetitive cartilage damage to the medial ankle (Valderrabano et al., 2006). Financially, osteoarthritis is associated with a number of direct (medication and health care) and indirect (days lost from work and benefits) costs (Chen, Gupte, Akhtar, Smith, & Cobb, 2012). The development of enhanced rehabilitation and preventative measures for lateral ankle sprains will have a knock-on effect to the prevalence and expenditure associated with posttraumatic ankle osteoarthritis.

2.5.1 Assessing Ankle Instability

Diagnosis of CAI is controversial in clinical practice and in academic literature. Traditionally pathologies are diagnosed with the use of clinical skills, imaging and questionnaires (Simon, Donahue, & Docherty, 2014). There is however no gold standard for diagnosing CAI (Tanen et al., 2014). Some studies have used clinical tests to define MAI and FAI. Functional ankle instability was described as having a negative talar tilt and anterior drawer tests along with the reported feeling of giving way. However this has been critiqued for not assessing within a weight-bearing position and due to the variation in assessment between clinicians (Monaghan et al., 2006).

Self-reported outcome instruments are used in most research studies to collect subjective information from individuals to determine the presence of ankle instability (Carcia, Martin, & Drouin, 2008). These have been used in a clinical and research setting (Hiller et al., 2006). These self-reported instruments can be either evaluative or discriminative. Evaluative instruments assess the effectiveness of the treatment and the injury outcome by measuring the change in injury status against time. In contrast, discriminative instrumentation is used to identify whether individuals present with FAI (Carcia et al., 2008).

There are currently a number of questionnaires in use in research and by clinicians to diagnose CAI, however, there is no consensus as to which is the gold standard (Wikstrom et al., 2009). Commonly used questionnaires include the Ankle Instability Instrument (AII), Chronic Ankle Instability Scale (CAIS), Ankle Joint Functional Assessment Tool (AJFAT), Cumberland Ankle Instability Tool (CAIT), Foot and Ankle Instability Questionnaire (FAIQ), Foot and Ankle Ability Measure (FAAM), Foot and Ankle Outcome Score (FAOS) and most recently the Identification of Functional Ankle Instability (IdFAI) questionnaire (Donahue, Simon, & Docherty, 2011; Tanen et al., 2014).

The AII is a discriminative questionnaire (Carcia et al., 2008) that consists of nine yes/no questions, six multiple choice questions and an open-ended question (Donahue et al., 2011). These questions can be divided into 3 categories - the severity of the initial sprain, history of ankle instability and instability in activities of daily life (Donahue et al., 2011). The guidelines set out in the IAC position statement stated CAI is indicated when individuals answer ‘yes’ to at least five yes/no questions (this must include question 1) (Gribble et al., 2014). A recent study found good reliability of the AII when used in conjunction with the CAIT (Donahue et al., 2011). This questionnaire gathers a large amount of information on the severity of the sprain and also the level of functionality and the perception of giving way. However, within the definition of CAI perceptions of weakness and pain are also thought to be important which this questionnaire does not cover. It also lacks definitions for the terms ‘giving way’ and ‘unstable’.

The CAIS is a 14-item questionnaire which covers disability, impairment, emotion and issues with participation (Donahue et al., 2011). Items are scored on a four-point Likert scale between 0 (worst score) and 4 (best score) higher scores indicate higher ankle stability (Donahue et al., 2011; Eechaute, Vaes, & Duquet, 2008). In comparison to the AII, this questionnaire does not provide as much information on the sprain severity.

The AJFAT is a 12 item questionnaire with 5 responses on a 48 point scale where higher scores indicate fewer symptoms and greater function (Hiller et al., 2006; Wikstrom et al., 2009). The responses involve comparing between ankles therefore not distinguishing bilateral from unilateral instability (Donahue et al., 2011; Ross, Guskiewicz, Gross, & Yu, 2008). For this reason, it may be an unsuitable method for determining whether CAI is present.

The CAIT is another example of a discriminative tool (Carcia et al., 2008). It was created to determine whether an individual has FAI and also to grade the severity of the instability (Hiller et al., 2006). Unlike the AJFAT and FAIQ this questionnaire asks individuals to individually grade both ankles instead of comparing to the contralateral ankle thus allowing to identify whether the individual has unilateral or bilateral instability (Donahue et al., 2011; Hiller et al., 2006; Tanen et al., 2014). The CAIT is a 9 item questionnaire, each answer is assigned a point value between 0 and 5 with a maximum score of 30 indicating the highest stability (Donahue et al., 2011; Marshall, McKee, & Murphy, 2009). The initial study suggested that individuals scoring 27 or lower were likely to have FAI (Hiller et al., 2006), however, this cut-off score was thought to be too high as athletes who had suffered from ankle sprains but had no residual symptoms were classified as suffering from CAI (Wright, Arnold, Ross, & Linens, 2014). This cut-off score was therefore revised and the IAC position statement now recommends a score of 24 or lower (Gribble et al., 2014). The CAIT has been found to have an 82.9% sensitivity and 74.7% specificity along with 0.96 intraclass correlation (Marshall et al., 2009).

The FAIQ is composed of yes/no questions therefore making it insensitive to the severity of instability (Hiller et al., 2006). These questions cover sensations of weakness, giving way during daily activity and injury (within past 3 months). Individuals must answer yes to certain questions and no to others. No studies have been carried out for reliability with this questionnaire (Donahue et al., 2011). Again, this questionnaire requires comparison to the other ankle sprain therefore not allowing for the possibility of bilateral CAI.

The FAAM is an evaluative questionnaire which was designed based on a previously used questionnaire called the foot and ankle disability index (Carcia et al., 2008; Donahue et al., 2011). There are two parts to this questionnaire: activities of daily living and sport (Wright et al., 2013a). The IAC recommends the use of this questionnaire for describing the level of disability with a cut-off score of 42 with less than 90% on the ADL scale and less than 80% on the sport scale (Gribble et al., 2014). It is, however, not suitable for determining the presence of CAI and the frequency of ankle sprains and severity of ankle sprains.

The FAOS is a 42 item questionnaire covering 5 areas - pain, other symptoms, sport and recreational function, foot and ankle related quality of life and activities of daily living (Donahue et al., 2011). This has been criticised for not including questions on the feeling of giving way and the recurrence of ankle sprains and for including items such as pain at night which are not specific to ankle instability therefore potentially jeopardizing the validity of this questionnaire (Eechaute et al., 2008). The IAC position statement recommended the FAOS for describing the level of disability with a score of less than 75% in three or more categories indicating instability (Gribble et al., 2013).

The IdFAI was most recently devised specifically to detect CAI in a clear and concise manner and be quick to administer (Simon et al., 2014). This was designed based on the CAIT and the AII (Simon, Donahue, & Docherty, 2012). This has been found to have an 89.6% accuracy and intraclass correlation of 0.92 (Simon et al., 2014). A study investigating the reliability of the IdFAI in 120 adults between the ages of 20-60 years found excellent levels of test-retest reliability using ICC of 0.978, 0.975, 0.961 and 0.922 for the 20-30 years, 30-40 years, 40-50 years and 50-60 years respectively (Gurav, Ganu, & Panhale, 2014). The IAC recommends its use with a cut-off score of 11 or more indicating CAI (Gribble et al., 2014). This questionnaire effectively combines the AII and the CAIT to produce a succinct questionnaire which covers the severity and the level of instability.

The IAC suggests predominately the use of the AII, CAIT or IdFAI for discriminative purposes and the FAAM or the FAOS for evaluative purposes where relevant to the research question (Gribble et al., 2013). This study will aim to distinguish between individuals with and without ankle instability. The IdFAI seems to be a valid and reliable measure to use combining the best elements of the AII and the CAIT and therefore will be used for classification purposes within this research.

2.5.2 Kinetic Chain

The initial kinetic chain concept is proposed by Franz Reuleaux and was initially related to engineering, however, this concept was translated across to human movement by Hans von Baeyer in 1933 at the International Orthopedic Congress and is now often applied within a rehabilitation context (Karandikar & Vargas, 2011). The kinetic chain principle states that a combination of successively arranged joints constitutes a complex unit and as such movement of one joint affects the movement of another (Karandikar & Vargas, 2011). It suggests that the body is a multi-linked system with for example the rectus femoris, hamstrings and gastrocnemius muscles crossing the hip, knee and ankles. It is suggested that movement of the trunk (which accounts for 35.5% body mass) will also have an impact on the motion of the hip and therefore knee and ankle (Kulas et al., 2008). With foot placement, small errors are thought to be corrected by the subtalar joint and larger errors are thought to be corrected at the hip joint, therefore, analysis of the full kinetic chain when analysing movement may provide greater detail of the whole movement pattern (Friel, McLean, Myers, & Caceres, 2006).

2.5.3 Feedforward and Feedback Strategies

Individuals with CAI have been reported to have insufficiencies in the feedforward and feedback strategies of motor control (Yen, Corkery, Donohoe, Grogan, & Wu, 2016). Feedforward control is suggested to describe actions occurring on identification of the beginning, and also includes the impending events or stimulus, whilst feedback control describes actions occurring in response to sensory detection of effects from the arrival of the event or stimulus to the system (Riemann & Lephart, 2002).

2.5.3.1 Feedforward Motor Control

Feedforward can be termed predictive or proactive – it is pre-planned and unchanged by peripheral feedback (Bastian, 2006). Feedforward motor control can be described as "the anticipatory actions that occur prior to sensory detection of a homeostatic disruption" (Riemann & Lephart, 2002). It is suggested that fast movements cannot exclusively utilise feedback control since biological feedback loops are inherently slow, thus the brain predicts sensory consequences based on efference copies of previously issued motor commands (Kawato, 1999).

2.5.3.2 Feedback Motor Control

Feedback motor control can be described as "the corrective response within the corresponding system after sensory detection" (Riemann & Lephart, 2002). Feedback strategies can be termed reactive whereby corrections are made to the movement based on information received (Bastian, 2006). This information is obtained via proprioceptive feedback (Brooks, 1983). Feedback is suggested to come from sensory afferents that are activated during the movement itself (Zewdie, Roy, Okuma, Yang, & Gorassini, 2014). The central nervous system receives information from three subsystems - the visual system, the vestibular system and the somatosensory system (Lephart, Pincivero, & Rozzi, 1998). The somatosensory system receives information from peripheral articular and musculotendinous receptors regarding changes in length and tension of the muscles and also information on joint position and motion. Mechanoreceptors detect when range of motion nears its limit and senses joint compression. It also provides protection at extreme range of motion. Mechanoreceptors are located within the skin, the musculotendinous unit, within the bone, the joint ligaments and the joint capsule (Lephart et al., 1998). There are four classifications of mechanoreceptors (type I-IV). A cadaveric study by Michelson & Hutchins (1995) found the ankle to contain a low number of type I mechanoreceptors (slow-adapting receptors with a low threshold), type II mechanoreceptors (dynamic, quick adapting receptors with a low threshold) were found in large quantities in all ligaments of the ankle along with type III (slow-adapting, dynamic receptors with a high threshold). Type I mechanoreceptors are thought to help facilitate postural sense, whereas type II, on the other hand, are postulated to sense initiation in joint movement. Type III mechanoreceptors are thought to be active during extremes of motion, therefore alerting the central nervous system to joint danger. Type IV mechanoreceptors are responsible for nociceptive sensation however none were found during the cadaveric study.

Two common examples of mechanoreceptors are Golgi tendon organs and muscle spindles (Martini, Nath, & Bartholomew, 2011). Golgi tendon organs are responsible for monitoring variations in muscle contractile forces (Stefanini & Marks, 2003). Golgi tendon organs are stimulated by tension within the tendon, excessive stimulation causes the contraction strength to be decreased (Martini et al., 2011). Muscle spindles are responsible for monitoring changes in muscle length and the rate that these changes occur (Needle et al., 2013). When rapid inversion occurs the muscle spindles in the peroneals are activated and a reflex contraction of the peroneus longus and brevis occurs to counteract this (Knight & Weimar, 2011b). Damage to these receptors is thought to decrease proprioceptive control of the ankle joint and predispose the ankle to FAI (Freeman et al., 1965).

Following injury, it is suggested that the normal reaction pattern of muscles is insufficient to protect the ankle joint from injury and as such a centralised feedforward neural adaption is implemented in order to help protect the joint using proximal and distal strategies (K. A. Webster, Pietrosimone, & Gribble, 2016). It is suggested that the damage to proprioceptors that occurs with the initial injury may also disrupt the proprioceptive feedback to the central nervous system and may be the reasoning for altered movement patterns following IC (Yen et al., 2016).

2.5.4 Ankle and Hip Strategy

Research has suggested that individuals utilise either an ankle or hip strategy (Horak & Nashner, 1986; Kuo & Zajac, 1993; Runge et al., 1999). It is suggested that for smaller perturbations or external stimuli an ankle strategy is utilised but for larger disturbances, a hip strategy is utilised (Kuo, 1995). The ankle strategy involves correction of the centre of gravity (to maintain above the base of support) by moving the body about the ankle joint. In comparison, the hip strategy utilises hip flexion and extension in order to maintain the centre of gravity within the base of support (Horak & Nashner, 1986). The hip strategy utilises more trunk rotation through the use of hip movements (Horak, Nashner, & Diener, 1990). The ankle strategy utilises an ankle-knee-hip muscle activation strategy whilst the hip strategy uses a primarily proximal hip muscle activation strategy (Horak et al., 1990).

Some research investigating kinematics and kinetics in individuals with CAI has reported a "top-down" or hip strategy to maintain balance (Hubbard et al., 2007; Tropp et al., 1985). It is suggested that proximal adaptations may enable normal kinetics to be observed when measuring force plate variables such as centre of pressure (Abdelraouf, Elhafez, & Abdel-Aziem, 2012). It remains unclear whether this is an attempt to maintain stability by locking movement of the unstable ankle joint or whether this is a compensatory strategy that is used to account for damage to the ankle joint. More thorough analysis combining trunk and detailed lower extremity kinematics is called for.

2.5.5 Limb Dominance

An epidemiological study found 48.2% of individuals to suffer from bilateral ankle sprains and 51.8% unilateral. Of the unilateral ankle sprains, injuries to the dominant leg were 2.4 times more likely (Yeung, Chan, So, & Yuan, 1994). This is proposed to be due to increased exposure to inversion forces in jumping and kicking. Comparatively a study of ankle sprains within English Premier League and Football League clubs observed no significant differences between the incidence of dominant and non-dominant ankle sprains (Woods, Hawkins, Hulse, & Hodson, 2003). Clinically comparison is drawn to the non-injured side to determine severity and criteria for return to play.

Inconsistencies exist in comparison studies involving control groups and instability groups. Some studies include those with injuries to the dominant and non-dominant leg and match this variable when comparing to the control group (Knight & Weimar, 2012; Koshino et al., 2014). Others discuss dominance for the control group but do not report this within the ankle instability group (Knight & Weimar, 2011b). Others simply match the side and ignore the involvement of dominance (Kipp & Palmieri-Smith, 2013). There are several different definitions of the dominant limb in the literature. It is suggested that the dominant limb is 'the leg used in order to manipulate an object or to lead out in movement' (M. Peters, 1988). A study investigating the link between self-reported limb dominance and observed dominant leg observed a 100% agreement with the criteria 'if you would shoot a ball on a target, which leg would you use' (van Melick, Meddeler, Hoogeboom, Nijhuis-van der Sanden, & van Cingel, 2017).

2.6 Preventative Measures for Ankle Sprains

Due to the high incidence of ankle sprains a number of preventative measures have been introduced (de Noronha, França, Haupenthal, & Nunes, 2013). Adopted intervention strategies have included external support in the form of ankle braces and taping methods, balance and coordination training, orthotics, footwear characteristics, strengthening and stretching (McKeon & Mattacola, 2008).

2.6.1 Taping and Bracing

Ankle taping and bracing are commonly used to prevent ankle sprains from occurring. These provide external support to the ankle, therefore, resisting active and passive inversion of the ankle (Shima et al., 2005). External supports have been reported to decrease ankle sprain re-injury rates by 50-70% (McKeon & Mattacola, 2008). A meta-analysis performed by Cordova, Ingersoll and LeBlanc, (2000a) examined a total of 253 cases across 19 studies comparing basketweave taping, lace-up braces and semi-rigid braces before and after exercise. Semi-rigid braces have been found to provide the greatest restriction in inversion followed by lace-up braces and then ankle taping. Ankle braces are often used due to their ease of application, re-usability and their cost-effectiveness when compared to ankle taping (Cordova, Dorrough, Kious, Ingersoll, & Merrick, 2007; Shima et al., 2005). Braces have, however, only been proven to reduce the incidence of recurrent ankle sprains (Verhagen & Bay, 2010) there is therefore doubt as to whether this method is a suitable preventative measure for all individuals who have not previously suffered an ankle sprain. Research has, however, observed more proximal adaptation to ankle bracing during two movements- turning sideways to catch a ball on one leg and turning to touch a target with their shoulder (Santos et al., 2004). They observed decreased trunk rotation and increased axial rotation of the knee in the braced condition therefore placing the ligaments and connective tissue under increased stress increasing the risk of knee injury. However, the study only observed 10 healthy participants.

Ankle taping is one of the most commonly used preventative measures for ankle sprains and has been found to decrease ankle sprains between two and fourfold when compared to other preventative measures (Cordova, Cardona, Ingersoll, & Sandrey, 2000b;

Verhagen & Bay, 2010). Its effectiveness, however, has been questioned as its level of support has been shown to decrease with time (Shima et al., 2005). One study found ankle taping to produce no significant differences in scores from a hopping test and the modified Star Excursion Balance Test, however, participants' perceptions of stability, confidence and reassurance were improved with a real and a placebo taping technique (Sawkins, Refshauge, Kilbreath, & Raymond, 2007).

Although taping and bracing has been found to have similar preventative effects in individuals with a history of ankle sprains, taping is thought to cost between 3-25 times the costs of bracing (McKeon & Mattacola, 2008). It has been postulated that the use of external ankle supports can lead to individuals developing weakness of surrounding ankle musculature and decreased neuromuscular function, therefore, leading to increased risk of injury (Cordova et al., 2000b). Cordova et al. (2000b) analysed electromyographic latency of the peroneus longus after sudden inversion without a brace, with a semi-rigid brace and a lace-up brace and found no difference in peroneal latency. Each participant was then assigned to a condition (control, active ankle or McDavid brace) and were required to wear the brace a minimum of 8 hours a day, 5 days a week for an 8-week period. Following this period, they found no changes in peroneal latency. However, another study by Shima et al. (2005) again analysed peroneal latency using surface electromyography in athletes with and without a history of ankle sprains. This was measured during a 25° inversion perturbation and found ankle taping and bracing to delay the peroneal reflex latency in hypermobile, injured and intact ankles.

It has been proposed that changes in ankle kinematics due to taping and bracing may cause changes in kinematics and energy absorption of the hip and knee joints (Cordova et al., 2010). Contrary to expected results, a study analysing straight line runs, 45°

sidesteps and 45° crossover cuts found ankle taping to provide a level of protection to the knee joint by decreasing internal rotation, varus moments and varus impulses (Stoffel et al., 2010). Another study analysed knee and ankle kinematics and ground reaction forces during a standardised drop landing with and without the use of a brace and found significantly increased knee flexion at initial ground contact which they associated with decreased anterior cruciate ligament loading (DiStefano et al., 2008).

2.6.2 Balance and Co-ordination Training

Some rehabilitation programmes implemented following ankle sprains tend to focus on addressing this proprioceptive deficit with balance and coordination training. Balance training is performed in weight bearing and is thought to improve mechanoreceptor function, re-establish the normal neuromuscular feedback loop and improve functional ability (V. M. Clark & Burden, 2005; Rozzi, Lephart, Sterner, & Kuligowski, 1999). Balance and coordination training often involves a single-limb stance and activities that challenge the individual's level of stability. They often utilise dynamic hopping exercises, balance boards or foam pads to create an unstable surface to further challenge the individual. (McKeon & Mattacola, 2008). McKeon et al. (2009) analysed kinematic measures for rearfoot inversion and eversion, shank rotation and the relationship between these during walking and running in participants with CAI. Participants were divided into a balance training group and a control group. The balance training group undertook 4 weeks of supervised training focusing on single limb stance exercises. They observed a significant decrease in shank (internal/external rotation)/rear foot (inversion/eversion) coupling variability in walking following the balance training suggesting improved stability (McKeon et al., 2009). Mohammadi (2007) analysed re-injury rates across a season in 80 first division male football players who had sustained ankle sprains. They were split into four groups; the first group followed a balance training programme using an ankle disk for 30 minutes each day, the second group followed a strength training programme for the evertors, the third group used an ankle brace and the fourth group were the control group. They found the incidence of ankle sprains in the balance training group to be significantly lower than the control group however no other significant differences were observed between groups. Although this looks promising, lateral ankle sprains were still sustained therefore further understanding of differences present in those with recurrent sprains may further strengthen this intervention.

2.6.3 Strengthening Within Rehabilitation Programmes

Strengthening has long been an important aspect of rehabilitation programmes for ankle inversion sprains (Wilkerson et al., 1997). Strengthening exercises performed within rehabilitation programmes often focus on the evertors (peroneals and extensor digitorum longus) and the dorsiflexors (tibialis anterior and extensor digitorum longus) (Holmes & Delahunt, 2009; Mohammadi, 2007). Eccentric contraction of the evertor muscles is known to resist ankle inversion and also support the lateral ligaments of the ankle, weakness of these muscles may contribute to recurrent ankle sprains (Caulfield, 2000). Increased muscle activation levels brought about during rehabilitation have been found to contribute to decreased ankle joint stiffness (Lin, Chen, & Lin, 2011). Muscular contraction has also been postulated to acutely affect the sensitivity of mechanoreceptors, therefore, linking strength and proprioception. One study found that performing strengthening exercises for the ankle improved joint position sense in inversion and plantarflexion in participants with FAI which they believed was due to increased gamma-efferent activity (Docherty, Moore, & Arnold, 1998).

2.6.4 Lack of Consensus

Although several preventative measures are proposed within the literature, a lack of consensus and a high epidemiology of lateral ankle sprains still exists resulting in a large cost implication and impacts on quality of life (Kerkhoffs et al., 2012; McGuine & Keene, 2006). A clear understanding of the injury mechanism along with biomechanical quantities is thought to be extremely important in the development of injury prevention strategies and protective equipment (Bahr & Krosshaug, 2005; Fong et al., 2012). This suggests that existing methods may be enhanced with increased knowledge of the mechanism and biomechanical quantities (Fong et al., 2012; Kristianslund et al., 2011).

2.7 Motion Analysis

Motion analysis can be two or three-dimensional (R. Li, Tian, Sclaroff, & Yang, 2010). Three-dimensional is considered to be the gold standard (Munro, Herrington, & Carolan, 2012). Optical 3-dimensional systems use infrared cameras to track the motion of markers creating a 3D trajectory of the path of each marker (Carse, Meadows, Bowers, & Rowe, 2013).

Three-dimensional motion analysis is widely used in research and also for clinical assessment to assist with clinical decision making and to evaluate the outcome of therapeutic interventions in those with disabilities (Groen, Geurts, Nienhuis, & Duysens, 2012). Positional data is recorded against time, enabling calculation of kinematic measures for example displacement, velocity and acceleration (Melton, Mullineaux, Mattacola, Mair, & Uhl, 2011). Motion analysis has also been used to identify potential risk factors for injury (K. E. Webster, McClelland, Wittwer, Tecklenburg, & Feller, 2010).

Human movement tracking can be either visual or non-visual (Zhou & Hu, 2008). Non-visual involves sensors being attached to the body to collect information on the movement that is occurring, these sensors can be magnetic, mechanical, microwave, inertial, acoustic or radio based (Zhou & Hu, 2008). Magnetic tracking technology reports the location of sensors within a magnetic field which is generated by a transmitter source. These can, however, be significantly affected by interference from metal or magnetic fields in the environment and electrical devices (Aminian & Najafi, 2004). Visual tracking uses markers which are attached to bony landmarks to model the movement of the underlying skeleton.

2.7.1 Markers

Intracortical bone pins with markers on top are the gold standard for analysing bone motion however these are highly invasive therefore skin-mounted markers tend to be favoured (Deschamps et al., 2011). These minimally affect the movement of the participant being tested and provide no discomfort to the participant (Deschamps et al., 2011). Skin-mounted markers used in motion analysis can be either active markers which consist of light-emitting diodes or passive retro-reflective markers (Aminian & Najafi, 2004). Active markers use strobing light emitters to uniquely identify each marker. These avoid marker occlusion but must be wired to a power unit (Culmer, Levesley, Mon-Williams, & Williams, 2009). The wires are therefore restricting to the participant and limit the degree of dynamic testing that can be performed. Passive retroreflective markers reflect infrared light from the camera system to calculate three-dimensional joint rotations (Janura et al., 1998; Poppe, 2007; K. E. Webster et al., 2010). These markers are not limited by rotation like active markers are, however, these markers must be labelled using computer software which can be time consuming and is a common source of error (Allard, Stokes, & Blanchi, 1995). Passive markers are wireless and therefore are very sensitive to marker occlusion, which is where a marker is hidden. This occurs when a marker, object or limb obscures another marker causing the system to mistakenly interchange the two (Culmer et al., 2009).

Marker size is an important consideration when analysing three-dimensional movement, it is important to ensure that the markers are large enough to be seen by the cameras and cover a suitable number of pixels but small enough for the area it is analysing so that they do not overlap or interfere with movement (Milner, 2008). Camera resolutions have improved significantly since the initial camera systems were introduced, this has meant that smaller markers can be used and placed closer together than had previously been possible, thus seeing the introduction of more detailed analysis of movement (Rankine, Long, Canseco, & Harris, 2008). Markers must be seen at all times by at least two cameras in order to identify its three-dimensional coordinates, if the marker is not visible to at least two cameras marker dropout will occur (Milner, 2008).

2.7.2 Camera Set Up

There are a number of things that should be considered prior to data capture with regard to camera set up which include capture volume, the sampling frequency and the lens options (Milner, 2008). The capture volume must be considered with reference to the activity that is being recorded, the volume must be large enough to capture the motion but not too large that the camera resolution is compromised (Milner, 2008). It is important to ensure that there are a suitable number of cameras for the capture volume and that these need to be spaced appropriately to limit dead space (part of the camera view that does not provide information for the data capture). This decreases camera resolution due to a decreased number of pixels covering the capture area (Milner, 2008). The correct sampling frequency must also be chosen to ensure that the data is suitably digitised and the Nyquist theorem states that the sampling frequency should be at least twice that of the highest signal frequency (Allard et al., 1995). The focal length (distance between focus and centre of a lens) and f-stop (amount of light that is allowed to pass through the lens) are often the modified settings when changing the lens options (Milner, 2008).

2.7.3 Calibration of Camera Systems

Calibration of motion analysis systems is a crucial step prior to beginning testing and this ensures that the coordinates of an image are correctly scaled and the two-dimensional image produced by each camera converted into three-dimensional coordinates (Milner, 2008; Robertson, Caldwell, Hamill, Kamen, & Whittlesey, 2013). The accurate calibration of the point of origin and the global coordinate system is extremely important as camera orientations and positions are defined from this information. Additionally, if additional items such as force plates are used, these too rely on this calibration and if these are inaccurate there will be an increase in joint moment error (Passmore & Sangeux, 2014).

Previously, a three-dimensional calibration cage has been used to calibrate camera systems, this used a single image to calibrate the system with known marker positions (Pribanić, Peharec, & Medved, 2009). However, this method has been superseded by wand calibration due to the ability to change the calibration volume size, the ease of storage and the decreased cost (Pribanić et al., 2009). The calibration now used is a two-stage process, the first stage is a static calibration this uses markers attached to a fixed structure, the coordinates of which are known (Robertson et al., 2013). This is often a rigid L-frame consisting of 4 markers and this frame defines the origin and orientation of the testing area (Ford, Myer, & Hewett, 2007; Milner, 2008). The second stage is a dynamic calibration, which calculates the lens focal length, the lens distortion maps and refines the camera positions on the software (Ford et al., 2007). The calibration wand has markers at known distances apart. The investigator is then required to walk through the calibration volume capturing the wand in as many orientations as possible (Pribanić et al., 2009). They must ensure that the whole volume is covered in a variety of wand orientations to ensure the three cardinal planes are calibrated to a high degree of accuracy (Milner, 2008). The dynamic calibration lasts between 60 and 120 seconds (Pribanić et al., 2009). To ensure high-quality data, it is advised that the system is recalibrated each day prior to data capture (Passmore & Sangeux, 2014).

2.7.4 Marker Placement

Two commonly used validated marker sets are the Helen Hayes marker set (Figure 2.3) (Davis, Õunpuu, Tyburski, & Gage, 1991; Manal, McClay, Stanhope, Richards, & Galinat, 2000) and the Cleveland Clinic marker set (Figure 2.4) (Manal et al., 2000).

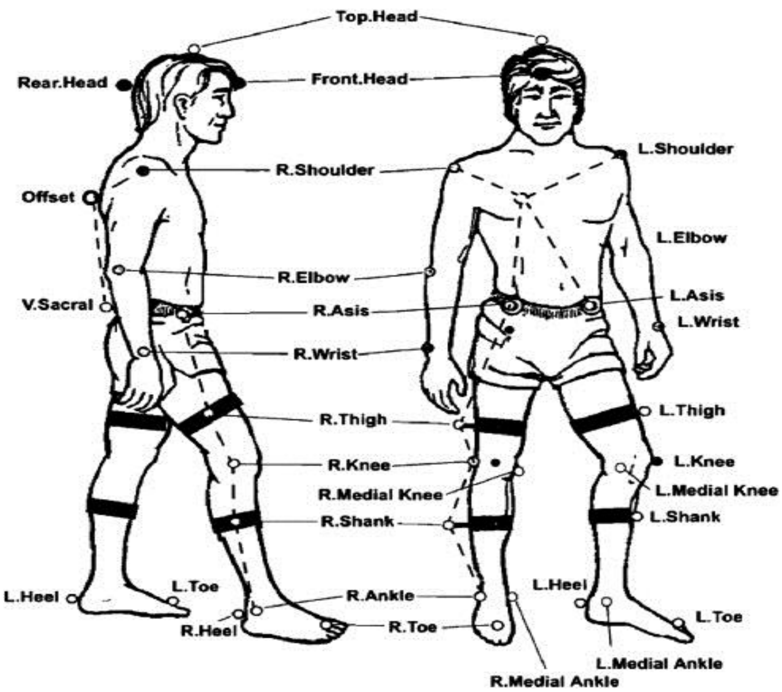


Figure 2.3 Helen Hayes Marker Set (Gallagher et al., 2009).

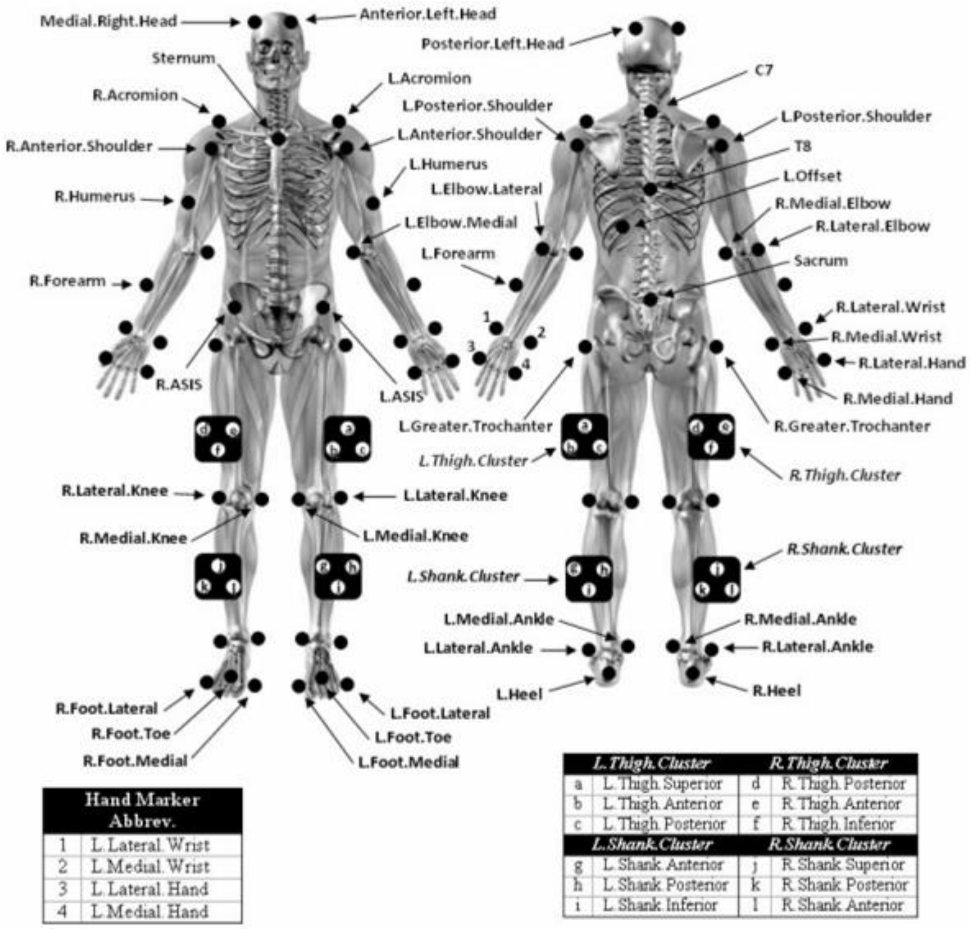


Figure 2.4 Cleveland Clinic Marker Set (MotionAnalysis, 2010)

Murali Kadaba developed the Helen Hayes marker set in 1985 whilst working in the Helen Hayes Hospital as a research scientist whilst the rivalling Cleveland Clinic marker set was later created by Kevin Campbell of the Cleveland Clinic Foundation (Sutherland, 2002). The Helen Hayes marker set is an anthropometric model whilst the Cleveland Clinic marker set is a cluster-based model (Long, Wang, & Harris, 2011). The Helen Hayes model involves the accurate placement of markers on bony landmarks and axes; this enables the bone and joint geometry to be calculated (Charlton, Tate, Smyth, & Roren, 2004). The Helen Hayes model can also utilise a wand-based marker set using wand markers on each segment to define the joint centres and segmental coordinate systems (Manal et al., 2000; Sutherland, 2002). The Cleveland Clinic marker set uses clusters placed on each segment to define the joint centre with medial and lateral markers for a static trial defining the flexion-extension axis, the medial and lateral markers are then removed and a dynamic trial is performed (Charlton et al., 2004; Radler et al., 2010). The Helen Hayes marker set is more simple to use than the Cleveland Clinic marker set and has been found to be more applicable to gait analysis in children (Sutherland, 2002). Radler et al. (2010), however, found less variability in kinematics of the transverse plane with the Cleveland Clinic marker set, thought to be due to less marker movement with three markers fixed to a rigid frame. Each of these marker sets provides a rigid foot to enable calculation of ankle kinematics, along with hip, knee, upper body and trunk kinematics.

The Calibrated Anatomical System Technique (CAST) has been implemented in recent studies (Sinclair & Bottoms, 2013). In this method, anatomical landmarks are located, calibration then occurs with the use of more technical markers on the participant's limbs (Cutti, Paolini, Troncossi, Cappello, & Davalli, 2005). This is thought to reduce movement artefact seen with other marker sets (Collins, Ghoussayni, Ewins, & Kent, 2009; Sinclair & Bottoms, 2013). This technique uses a static calibration of all anatomical landmarks and a dynamic calibration through full range of motion (Leardini, Chiari, Della Croce, & Cappozzo, 2005). The optimised lower limb gait analysis technique (OLGA) uses the same method with static and dynamic trials, however, uses an anthropometric model as opposed to the cluster based model (Groen et al., 2012).

When modelling the trunk the simplest method is to consider this as one rigid segment. Plug in models adopt this methodology (Preuss & Popovic, 2010). Practically this is the easiest method to adopt due to the number of markers required though this may cause intersegmental movement to be missed (Preuss & Popovic, 2010). The trunk has 33 vertebrae however modelling this has proven to be challenging due to the size of each vertebrae not allowing for multiple marker placement on each segment (Konz et al., 2006). As such many models group vertebrae together to produce segments for analysis (Konz et al., 2006). Although some research suggests that this may produce a more accurate representation of trunk kinematics (Preuss & Popovic, 2010) this is at the expense of time both in terms of data collection and data analysis and clinically this may prove to be more challenging to implement.

2.7.5 Multi-Segmental Foot Models

Traditionally, research analysing the motion of the foot and the ankle has considered the foot as one single rigid segment (Bishop, Paul, & Thewlis, 2012; Kidder et al., 1996). The foot is composed of multiple bones and joints with a number of complex interactions (Okita, Meyers, Challis, & Sharkey, 2009) therefore this excludes motion between and within the different segments providing inadequate information on the biomechanics of the foot (Stebbins et al., 2006). More recently, the use of multi-segmental foot models which divide the foot into several segments which are treated as separate rigid bodies have been introduced and utilised (Carson, Harrington, Thompson, O'Connor, & Theologis, 2001; Deschamps et al., 2011; Okita et al., 2009). This development has allowed for a greater understanding of the function of the foot and ankle (Bishop et al., 2012). A number of marker sets have been created to analyse the biomechanics of the foot, which demonstrate substantial variations in the number of segments, marker placement, axes definitions and the mathematical interpretation of motion, thus limiting the comparability of results (Carson et al., 2001; Seo et al., 2014; Simon et al., 2006). The most commonly used markers sets for analysis of clinical populations include the Heidelberg Foot Measurement Method (HFMM), the Milwaukee Foot Model (MFM), the Oxford Foot Model (OFM), the Leardini Foot Model (LFM) and the Ghent Foot Model. Each of these marker sets has had some degree of validation and each varies in comparison to the others. For example, the number of markers is different in each of the models. The HFMM consists of 12 markers, the MFM 11, the OFM 13, the LFM 16 (Seo et al., 2014) and the Ghent 25 (De Mits et al., 2012). More markers enable more thorough analysis of the motion of the foot, however, with an increased number of markers, there will be an increase in the likelihood of marker placement error (Seo et al., 2014).

The foot is often split into three segments: the hindfoot, forefoot and hallux. Euler angles and the joint coordinate system are most commonly used to define the relationships between these segments (Baker & Robb, 2006; Rankine et al., 2008; Saraswat, MacWilliams, & Davis, 2012). Euler angles refer to the rotations about fixed or intermediate axis of a rigid body in relation to another and within the body, these generally assess the movement of the distal segment relative to the proximal. The order of axis selection is critical to determine the magnitude of rotation (Rankine et al., 2008). The Heidelberg Foot Measurement Method however is named a method as opposed to a model as it doesn't use formal segment definitions (Deschamps et al., 2011)(Figure 2.5) instead it uses projection angles between functional segments (Rankine et al., 2008). Projection angles were defined by Simon et al. (2006) as: "the angle between two vectors in the perspective view along the axis of rotation". The HFMM reports on 12 motions, several of which are novel angles that they believed to be clinically relevant, such as medial arch angle which is thought to be important for several foot deformities (Rankine et al., 2008; Simon et al., 2006). It also introduced the use of a heel alignment device which allows for measurement of the differences between loaded and unloaded calcaneal positions and for biomechanical analysis of foot deformities. The device uses external manipulation to place the foot into a neutral position (Rankine et al., 2008). The number of angles this method reports is extremely useful in foot analysis however the use of the projection angle method as opposed to Euler angles has seen some critique for use in

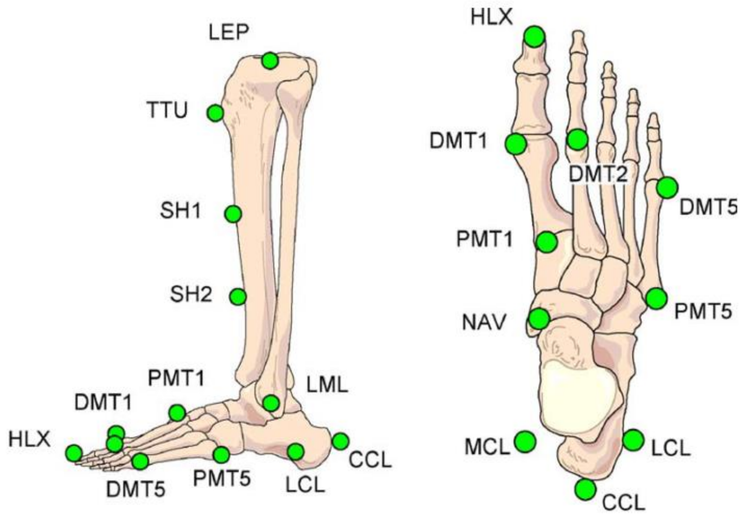
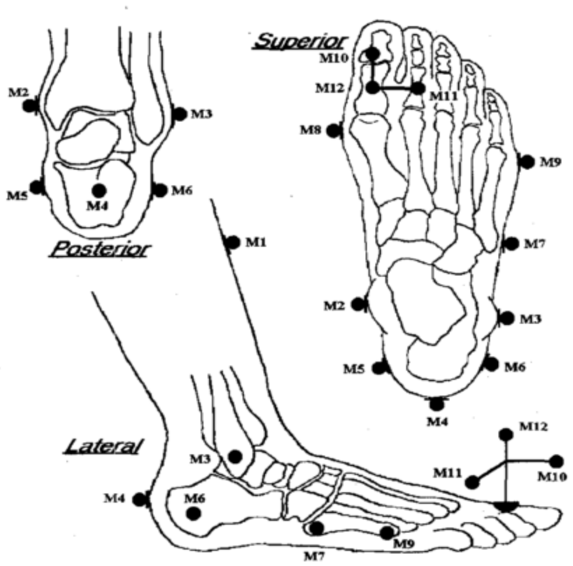


Figure 2.5 Marker placement for the Heidelberg

Foot Measurement Method (Simon et al., 2006)

research. As these projection angles are more difficult to interpret by clinicians and since the derived lines are not constrained to adjoining segments, segment-based models may be preferable (Baker & Robb, 2006).

The Milwaukee Foot Model (Figure 2.6) comprises the tibia, hindfoot, forefoot and hallux in a four-segment model. This model utilises x-rays in weight-bearing to ensure accurate placement over bone anatomy (Long, Eastwood, Graf, Smith, & Harris, 2010). This improves its use clinically for use with foot deformities however, there are cost and equipment implications for its use. The Milwaukee Foot Model was designed for use on the ageing foot but has been used by a number of other studies using different sample populations (Bishop et al., 2012). It does not have the ability to report movement between the hindfoot and midfoot and midfoot and forefoot segments and as such this and the requirement for radiographic images are key limitations of its use (Novak, Mayich, Perry, Daniels, & Brodsky, 2014).



M1	Medial surface of anterior tibia
M2	Medial malleolus
M3	Lateral malleolus
M4	Tuberosity of calcaneus
M5	Medial calcaneus
M6	Lateral calcaneus
M7	Tuberosity of 5th metatarsal (lateral)
M8	Medial head of 1st metatarsal
M9	Lateral head of 5th metatarsal
M10	Anteriorly directed hallux
M11	Laterally directed hallux
M12	Superiorly directed hallux

Figure 2.6 Marker placement for the Milwaukee Foot Model (Kidder, Abuzzahab, Harris, & Johnson, 1996)

The Oxford Foot Model (Figure 2.7) also consists of 4 segments- the hindfoot, forefoot, hallux and shank (Rankine et al., 2008). The initial Oxford Foot Model was proposed by Carson et al. (2001) however a paper by Stebbins et al. (2006) proposed a number of changes aiming to increase the repeatability of the marker set which now forms the accepted model (Wright, Arnold, Coffey, & Pidcoe, 2011). These changes included redefining the tibial segment using the conventional knee joint centre, altering the hindfoot segment so that it was independent of neighbouring segments, placement of the proximal marker on the first metatarsal was also modified to sit medially to the extensor hallucis longus tendon (Stebbins et al., 2006). This model is thought to illustrate some degree of external validity, however like the Milwaukee Foot Model, the Oxford Foot Model has been used for populations other than that it was designed for. The model was designed specifically for the use with paediatrics but a number of studies have used the model for adult populations (Bishop et al., 2012). Some criticism has also been raised with the marker set not accounting for motion between the hindfoot and midfoot and midfoot and forefoot which is a main limitation of this model (Novak et al., 2014).

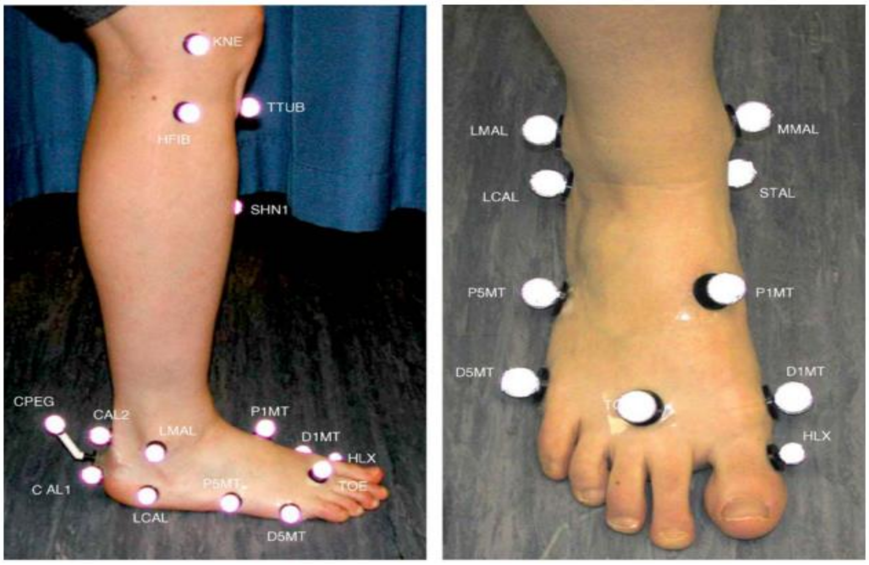


Figure 2.7 Marker placement for the modified Oxford Foot Model

(Stebbins, Harrington, Thompson, Zavatsky, & Theologis, 2006)

The LFM (Figure 2.8) is a five segment foot model which tracks the shank (tibia and fibula), the foot (including all bones), the calcaneus, the mid-foot (navicular and cuneiforms), and the metatarsus (five metatarsals) (Leardini et al., 2007). The LFM boasts the inclusion of a mid-foot segment allowing for further detailed analysis of the intricacies of foot movement (Powell, Williams, & Butler, 2013). It is suggested that although increasing the number of markers enables more precise analysis of segmental motion it also increases variability in kinematic data (Kim et al., 2018). A study comparing the HFM, OFM and also two other foot models (duPont and Utah) found high standard deviations in hindfoot and forefoot plantarflexion and dorsiflexion angles (Nicholson et al., 2018).

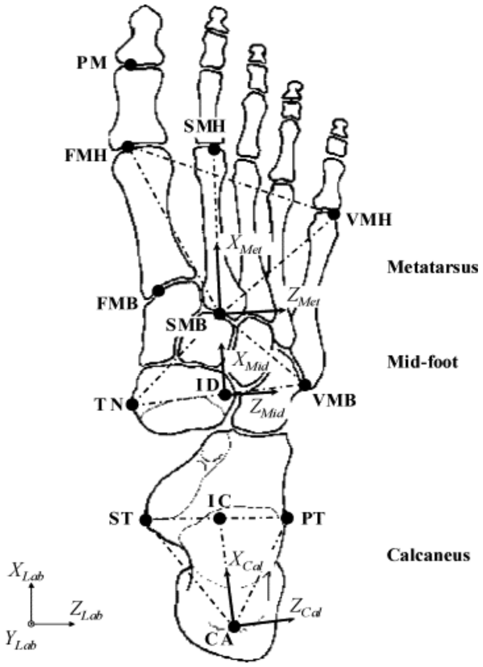


Figure 2.8 Marker placement for the Leardini Foot Model (Leardini et al., 2007).

The Ghent Foot Model (Figure 2.9) is a 6-segment foot model consisting of the lower leg, hindfoot, midfoot, medial forefoot, lateral forefoot and hallux segments (De Mits et al., 2012). Angles are referenced to a static reference standing position. This model, although it has been used in recent research of CAI populations, lacks data on reliability and validity in foot deformities (Novak et al., 2014). This also has a high number of markers which may pose an issue in more dynamic movement trials.

Figure 2.9 Marker placement for the Ghent Foot Model (De Mits et al., 2012)

The increased information that is provided by multi-segmental foot marker sets is necessary for the research into functional disability, however, there are two main problems with their use: the difficulty in overcoming skin motion artefact and the reproducibility of marker placement (Seo et al., 2014). As with full body marker sets the use of foot models relies on accurate placement of markers over anatomical landmarks. This is extremely difficult and with foot deformities it can be impossible (Saraswat, MacWilliams, Davis, & D'Astous, 2013). The proximity of marker placement on the foot amplifies errors in angular calculations in comparison with typical spacing over long bones. The reliability has also come into question due to the impact of skin movement artefact (Saraswat et al., 2012). One study used roentgen photography to analyse the movement of foot markers in relation to bone. The largest movements were found with the more proximally placed markers over the medial malleolus, calcaneus and navicular. The highest marker movement they reported was 4.3 mm (Tranberg & Karlsson, 1998). There is a need for further repeatability trials for the use of foot models within research as this is currently lacking (Deschamps et al., 2012).

2.7.6 Variability and Limitations of Motion Analysis

Variations in motion analysis data can be either intrinsic or extrinsic. Intrinsic factors, such as natural variation can only be measured and managed. Natural variability includes factors such as walking speed, height and the age of the participant which cannot be modified (Schwartz, Trost, & Wervey, 2004). Extrinsic factors on the other hand cover experimental errors which can be addressed to improve the quality of a study (Schwartz et al., 2004). There are three main sources of experimental error in motion analysis trials; the participant, the examiner and the measurement system (Gorton III, Hebert, & Gannotti, 2009).

During testing, participants may naturally alter mechanics creating variability. Other factors such as locomotion velocity have been found to create variability in data (Gorton III et al., 2009). Examiner error most often arises from marker placement variation which has been postulated to account for over 90% of the variability in motion analysis trials (Gorton III et al., 2009). This is due to the inconsistency and inaccuracy of marker placement on the correct anatomical landmark (K. E. Webster et al., 2010). It is essential that markers are accurately placed over relevant bony landmarks as these are then used for calculations of joint centres and the joint axis (Kainz, Carty, Modenese, Boyd, & Lloyd, 2015). Marker error also arises from soft tissue artefact, which can be due to motion of the skin, muscle and other soft tissue or due to movement over the bone (Collins et al., 2009; Fedie, Carlstedt, Willson, & Kernozek, 2010). The transverse plane is particularly prone to marker placement errors (K. E. Webster et al., 2010). In order to decrease this error, it is suggested that standardized protocols be designed and clear descriptions of marker placement outlined in methodology, where possible using single testers is also advised (Gorton III et al., 2009). Errors within the measurement system can arise from incorrect or inappropriate setup. For example, inappropriate spatial resolution or sampling speeds. Alternatively, errors may arise from the calculations within the system such as with regression-based joint centres (Schwartz et al., 2004). This error can be reduced with thorough configuration and calibration (Gorton III et al., 2009).

Soft tissue artefact refers to an error produced by movement of the skin mounted marker in relation to the bony prominence it is representing. This artefact is known to affect estimations of joint centre and rotation axis (Cerveri, Pedotti, & Ferrigno, 2005). This error varies based on individual characteristics, marker placements and the type of activity being analysed (A. Peters, Galna, Sangeux, Morris, & Baker, 2010). Soft tissue artefact most often occurs in the areas closest to the joints due to muscle contraction, inertial effects and skin deformation. It is very difficult to filter soft tissue artefact due to the similarities in frequency to bone movement (Leardini et al., 2005). Another possible source of error is within the estimations of joint centres using equations based on cadaveric specimens, this may result in error in the kinematic variables exported (Kainz et al., 2015). Plug in gait models have inherent issues with estimation of hip joint centre positions and also with defining the coronal plane of the femur producing artefact in hip rotation therefore caution should be taken in interpretation of these findings (Baker, Leboeuf, Reay, & Sangeux, 2018).

2.8 Electromyography

Electromyography is the most commonly used method for measuring muscle activation during exercise clinically and in research studies (Rainoldi, Melchiorri, & Caruso, 2004; Soderberg & Knutson, 2000). It is used to measure the electrical activity of skeletal muscles and also a representation of motor neuron outflow in the spinal cord to the muscles (Türker, 1993). It is therefore used in the study of muscle functional anatomy, motor unit firing and recruitment characteristics, biofeedback, neuron excitability and can be related to muscle force development and reflex connections (Perry-Rana, Housh, Johnson, Bull, & Cramer, 2003; Türker, 1993).

There are 2 types of electrode; intramuscular and surface (Okubo et al., 2010). Intramuscular electrodes can be either needle or wire-based, these are inserted directly into the deep or smaller muscles (Hug, 2011; Türker, 1993). Surface electrodes (sEMG) are active (built-in pre-amplification) or passive - the main difference being that passive are affected by changes in skin resistance (Türker, 1993). Surface electrodes are generally favoured due to their availability, ease of application and minimal patient discomfort (Soderberg & Knutson, 2000). They are, however, inappropriate for recording muscle activity of deep muscles (Murley, Menz, Landorf, & Bird, 2010). Surface electrodes are favoured for isotonic movements as needle electrodes are likely to displace during muscular contraction causing pain and damage to the muscles (Soderberg & Knutson, 2000).

Electrodes are most commonly used in bipolar configurations recording the potential between two electrodes on the muscle. Recordings can also be monopolar using one electrode to detect sEMG signal, however, this is less common as more noise is picked up from the vicinity and it is less muscle specific (Beck, DeFreitas, & Stock, 2011; Türker, 1993). A ground or reference electrode is also placed on a bony prominence to minimise noise within electromyographic recordings (Türker, 1993).

2.8.1 Electrode Placements

The correct placement of electrodes is essential with a displacement of just 1 cm between two measurements creating variations of up to 200% in estimates of amplitude (Rainoldi et al., 2004). SENIAM (2004) (Surface Electromyography for the Non-Invasive Assessment of Muscles) has developed recommendations that are now widely used within research for the recording of surface EMG outlining the electrode location, its orientation, the starting position and clinical tests for 30 individual muscles.

The electrical activity picked up from muscles other than that of the muscle under investigation is called crosstalk (Campanini et al., 2007). To reduce crosstalk, it is recommended that smaller electrodes are used for smaller muscles and a minimal electrode distance is implemented as surface electrodes are non-selective (O'Sullivan, Smith, & Sainsbury, 2010; Soderberg & Knutson, 2000). Mechanical artefact noise should also be minimised by using the shortest possible leads and active electrodes if possible (Türker, 1993). It is also recommended that the area for electrode placement be shaved and cleaned to remove dead skin and decrease electrical resistance between the electrode and the muscle (Distefano, Blackburn, Marshall, & Padua, 2009; Türker, 1993)

2.8.2 Normalisation of Electromyography

Electromyography is influenced by a number of factors: tissue characteristics, physiological crosstalk, location of the electrode, external noise and the electrode and amplifiers (Konrad, 2005). To reduce signal variation, sEMG recordings must be normalised (Albertus-Kajee, Tücker, Derman, & Lambert, 2010). It also enables comparison of data between tests, participants and studies. Without normalisation the significance of a study is limited (Boudreau et al., 2009; Ebben et al., 2009). Normalisation of data normally involves converting from millivolts to a percentage of a reference task, this improves absolute reliability (Ball & Scurr, 2010). The reference task used is generally a maximal voluntary contraction which can be either isometric or dynamic however recent research has introduced submaximal voluntary contractions (D. R. Clark, Lambert, & Hunter, 2012). The use of dynamic maximum voluntary contractions has been criticised for not maximally activating all muscles under investigation. Maximal voluntary isometric contractions (MVIC) have been found to be the most reliable method for normalisation of EMG data when compared to mean dynamic and peak dynamic normalisation methods (Bolgla & Uhl, 2007). The problem with MVICs arises with ensuring the contraction that the participant performs is as close to maximal as possible. This is highly dependent upon participant motivation levels (Albertus-Kajee et al., 2010) as a result the use of MVICs often produces supramaximal sEMG readings for submaximal dynamic tasks. This is potentially due to changes in muscle lengths during dynamic movements and motor unit synchronisation and increased superposition of electrical activity during dynamic movements (Konrad, 2005). There is controversy within research regarding the duration and number of repetitions that should be performed with MVICs though the standard procedure is three trials of 5-second duration (Soderberg & Knutson, 2000). A limitation of MVICs arises when working with clinical or unhealthy populations whereby producing a true maximal contraction is not possible (Konrad, 2005).

Other methods include normalisation of sEMG to the peak or mean of the dynamic movement under analysis (Burden & Bartlett, 1999). These have been reported to show good within participant and within day reliability however reliability of testing the same individual across days has been found to be less reliable (Halaki & Ginn, 2012). This method of normalisation compares to the task maximum and as such loses muscle innervation ratios and muscle activity levels cannot be compared between individuals, muscles or tasks; instead this method would be used to compare patterns of muscle activation (Halaki & Ginn, 2012). Mean and peak normalisation using the task under investigation is recommended when using symptomatic populations due to the potential inability to produce a maximal contraction due to pain or injury (Bolgla & Uhl, 2007).

2.9 Statistical Analysis

2.9.1 Statistical Parametric Mapping

Data is often reported using peaks or means or reported at specific time points. Biomechanical data is one dimensional (1D) (time and kinematic or force trajectories) therefore reporting reduced discrete data has the potential to result in focus bias or missing potential significance or trends during different phases of a movement (Pataky, Robinson, & Vanrenterghem, 2013). Statistical parametric mapping (SPM) is a concept introduced to biomechanics from brain research (Friston et al., 1994) which enables full curve analysis across the entirety of a movement (Pataky et al., 2013). This is suited to biomechanical data as this is temporally smooth in nature and can be temporally bound (Pataky, 2010). Statistical parametric mapping uses a concept called random field theory in order to control for the multiple comparisons being performed, this is in the place of a Bonferroni correction which is deemed to be overly conservative (Pataky, 2010). Statistical parametric mapping first estimates the smoothness of the residuals, then uses random field theory to determine a critical test statistic that retained a family-wise error of $\alpha = 0.05$, lastly, the probability that suprathreshold clusters could have been produced by chance is calculated (Pataky et al., 2013). Data is time normalised to 101-time nodes (Pataky et al., 2013). A limitation of SPM is the need for temporal registration of datasets potentially causing potential significance to not be highlighted appropriately (Pataky, 2010). Although this method raises a potential concern for significant differences to be missed this method may be a suitable method to analyse differences in movement patterns between individuals with and without CAI.

2.9.2 Time Series Analysis Using Confidence Intervals

Previously 1D confidence intervals have been used with 0D randomness models and as such have been deemed invalid (Pataky, Vanrenterghem, & Robinson, 2015). However, 1D bootstrap confidence intervals have been used and found to be a viable option for gait (Duhamel et al., 2004). This method involves complex computation using the mean, sample size, standard deviation, alpha and Gaussian (Duhamel et al., 2004). Although both methods have been deemed to be suitable for biomechanical analysis, comparison between SPM and time series analysis using confidence intervals concluded SPM to be the most suitable method for analysis of 1D data. This is due to increased generalisability of probabilistic conclusions (with the use of hypothesis testing techniques) and the ability to present results in a more consistent manner aiding interpretation of findings (Pataky et al., 2015).

2.10 Simulated Ankle Sprains

Tilt platforms are often used in research into CAI, they are designed to simulate the ankle sprain mechanism within injury free range (Fong et al., 2009a). Tilt platforms utilise trap doors which when released cause the participant to fall into a restricted position (Hopkins, McLoda, & McCaw, 2007). Previously used tilt platforms have faced criticism for only simulating the eversion/inversion aspect of ankle sprains (Chan, Fong, Yung, Fung, & Chan, 2008). Tilt platforms should replicate plantarflexion with inversion to suitably stress the ATFL (Eechaute, Vaes, Duquet, & Van Gheluwe, 2007; Mitchell, Dyson, Hale, & Abraham, 2008).

Some questions have been raised as to the validity of the tilt platform, specifically with the stationary tilt platform as there is thought to be a large anticipatory response (Hopkins et al., 2007). Tilt platforms can have one tilting platform which allows a set limb to be tilted or two tilting platforms where either side could be tilted, decreasing the participant's anticipatory response (Mitchell et al., 2008). Validity is also questioned with this model as sprains very rarely occur whilst weight is equally distributed across both limbs and instead occur when landing or running (Knight & Weimar, 2011a). During gait, muscle spindle sensitivity is increased due to increased muscle activity prior to and during the early stance phases of gait. This is thought to result in increased joint stiffness and a decreased reaction time, suggesting static results to be less ecologically valid than dynamic (Hopkins et al., 2007).

In order to better replicate ankle sprains in landing, the outer sole has been developed based on a previous study (Ubell, Boylan, Ashton-Miller, & Wojtys, 2003) which used a fulcrum to force the ankle into inversion to measure ankle brace effectiveness (Knight & Weimar, 2011a). The outer sole is a detachable fulcrum placed 20 mm from the medial border of the outer sole with metal attached to the lateral border (Knight & Weimar, 2011b, 2012). This is thought to produce inversion speeds similar to those reported during actual lateral ankle sprains (Knight & Weimar, 2012). This model has had little testing for reliability and validity (Knight & Weimar, 2011b). A study analysed the time to maximal inversion and the mean inversion speed and found high reliability between healthy and injured participants and greater reliability than tilt platforms for time to maximal inversion (Knight & Weimar, 2012). The ecological validity of this mechanism is questionable due to the height of the fulcrum therefore only replicating landing on another person's foot. Questions can also be raised over whether the high fulcrum would impact the protective mechanism for ankle sprains and whether the support limb would respond in the same way. Use of the tilt platform and outersole method may not be the most suitable for testing the effectiveness or for the design of new preventative measures for lateral ankle sprains due to the poor of validity and reliability - instead further biomechanical research is warranted.

2.11 Analysis of Ankle Sprains During Laboratory Testing and Sporting Competition

In some rare instances, video analysis has been carried out for ankle sprains in the laboratory or in sporting competition (Fong et al., 2009a; Kristianslund et al., 2011; Mok et al., 2011). One study involved participants running forward for 6 m then making a rapid left turn; this resulted in an accidental lateral ankle sprain from which biomechanics were subsequently analysed. However, digitisation was only performed for the tibial tuberosity, lateral malleolus, posterior shank, distal posterior shank, proximal heel, distal heel and toe tip. The ankle reached 48° inversion and 10° of internal rotation at the point of injury, this maximum position was reached at 0.20 seconds from FS (Fong et al., 2009a). Another study asked participants to use a sidestep cutting technique to move around a static defender resulting in a participant sustaining an accidental lateral ankle sprain; markers were placed on the legs, arms and torso, however, no results were obtained for these points instead the ankle was again the focus of the study. A sudden increase in inversion and internal rotation was observed between 130 and 180 ms following IC and an attempt to unload the foot 80ms after IC (Kristianslund et al., 2011). A significantly increased inversion velocity was also observed at 559°/s in the sprain trial vs 166 and 221 in the previous control trials (Kristianslund et al., 2011). Two ankle sprains sustained in the Beijing Olympic Games were analysed from televised video recordings focusing purely upon the ankle joint kinematics (Mok et al., 2011). The first lateral ankle sprain was sustained during take-off in a high jump event and the second was sustained during a field hockey match whilst the player was running under pressure. The maximum inversion angle for both case studies was found to occur 0.08 seconds after IC and at a velocity of 1752 °/s for the high jump injury and 1397 °/s for the hockey injury (Mok et al., 2011). Of these ankle sprains, no data has been reported for movement of the body superior to the tibial tuberosity. Of note again is the high-velocity inversion that is seen to occur within the injury case studies - this may be of value when investigating potential differences between healthy individuals and those with ankle instability.

2.12 Walking and Ankle Instability

Walking is one of the most basic and most utilised human movements. An epidemiological study of ankle sprain admissions into Accident and Emergency in the United States found 49.3% of ankle sprains occurred within athletic activity (Waterman et al., 2010). This leaves over 50% that occurred during activities of daily life. Of these a fall from the stairs accounted for 26.6%, however, a stumble at ground level accounted for 6.7% (Waterman et al., 2010). A review of ankle sprains in an Emergency Department in the south of England across a 7 month period found ankle sprains during walking to account for 12.2% of the analysed ankle sprains (Al Bimani et al., 2018).

Walking is commonly broken down into phases of the gait cycle. The period where the foot is in contact with the ground is referred to as the stance phase. This begins and ends with both feet on the ground (double-limb support). The remainder of the cycle is known as the swing phase of gait this is from toe-off to heel strike (HS) as the limb is swung forward (single-limb support) (Pirker & Katzenschlager, 2017) (Figure 2.10).

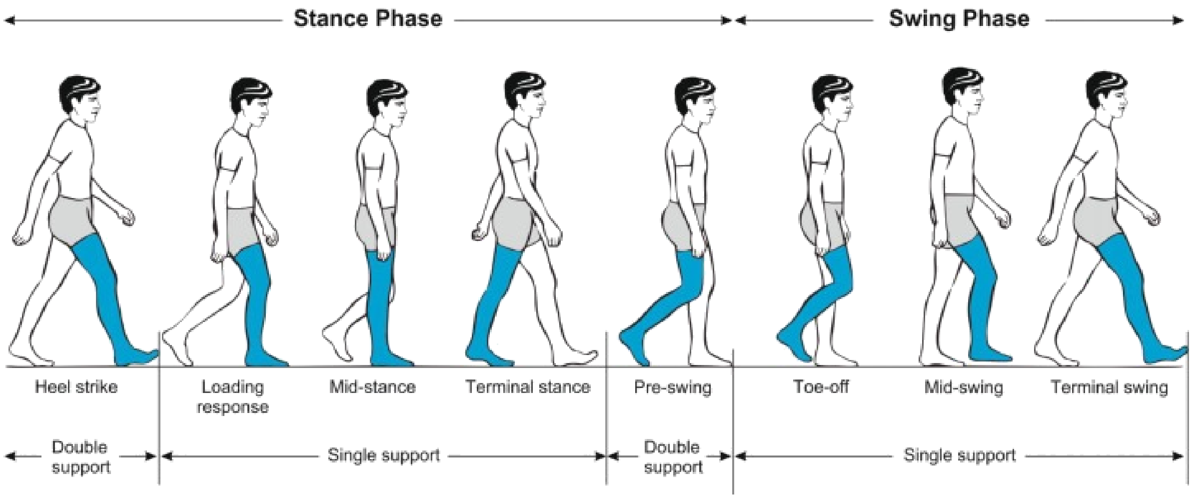


Figure 2.10 Phases of gait (Pirker & Katzenschlager, 2017)

Differences in postural control, kinematics, muscular activation and muscle onset times may predispose individuals to further episodes of giving way and recurrent sprains. Increased ankle inversion observed when analysing frontal plane kinematics during walking has also been found to correspond to greater ankle inversion during more sport-specific movements such as jump landing (Donovan & Feger, 2017). As previously suggested, repetitive loading of abnormal kinematic movement patterns has been linked to increased cartilage damage and as such the development of osteoarthritis within the ankle joint complex (Valderrabano et al., 2006), therefore thorough investigation into kinematics and muscle activation patterns is prudent.

Some research to date has investigated the kinematic differences in walking gait between individuals with ankle instability and healthy control participants (Chinn, Dicharry, & Hertel, 2013; De Ridder et al., 2013; Delahunt et al., 2006a; Drewes et al., 2009a; Herb et al., 2014; Monaghan et al., 2006; Terada et al., 2015; Wright, Arnold, Ross, & Pidcoe, 2013b). Others have investigated electromyography (Delahunt et al., 2006a; Feger, Donovan, Hart, & Hertel, 2015; Hopkins, Coglianese, Glasgow, Reese, & Seeley, 2012; Koldenhoven, Feger, Fraser, Saliba, & Hertel, 2016; Lin et al., 2011; Santilli et al., 2005) and some kinetics (Hopkins et al., 2012; Koldenhoven et al., 2016; Monaghan et al., 2006; Nyska et al., 2003). Lateral ankle sprains are not just prominent in sport they also affect the general population during activities such as walking. As such it is necessary to have a good understanding of biomechanics to better inform preventative and rehabilitation strategies.

2.12.1 Electromyographic Findings in Walking and CAI

Existing literature has documented sEMG during walking in individuals with CAI and with healthy controls for the gluteus medius (Feger et al., 2015; Koldenhoven et al., 2016), tibialis anterior (Delahunt et al., 2006a; Feger et al., 2015; Hopkins et al., 2012; Koldenhoven et al., 2016; Koldenhoven, Feger, Fraser, & Hertel, 2018) gastrocnemius (Feger et al., 2015; Koldenhoven et al., 2016), peroneus longus (Delahunt et al., 2006a; Feger et al., 2015; Hopkins et al., 2012; Koldenhoven et al., 2016; Koldenhoven et al., 2018; Santilli et al., 2005), rectus femoris (Delahunt et al., 2006a; Feger et al., 2015), biceps femoris (Feger et al., 2015) and soleus (Delahunt et al., 2006a). Although several studies have investigated sEMG during walking, differing methodologies exist with regard to reported variables, and processing and normalisation methods. Although Koldenhoven et al. (2016), Hopkins et al. (2012) and Delahunt et al. (2006a) all reported muscle activation during walking, the methods and time periods of reporting varied. With Koldenhoven et al. (2016) using the area under the sEMG RMS curve for 100 ms pre IC and 200 ms post IC time periods, Hopkins et al. (2012) RMS the data across a 50 ms time window and time normalised the data across the stance phase and Delahunt et al. (2006a) calculating integral EMG for 200 ms pre and 200 ms post HS. Different methods for normalisation were also used with the use of baseline sEMG during standing (Hopkins et al., 2012; Koldenhoven et al., 2016) and peak EMG (from the mean of 10 records) (Delahunt et al., 2006a) used to enable comparison between groups. The use of baseline standing as a method for normalisation is not a commonly used method within sEMG but was justified as the most stable and consistent reference value (Hopkins et al., 2012). Feger et al. (2015) reported activation time and duration at HS and also muscle activation for 100 ms pre and 200 ms post. They again RMS the data and normalised to quiet standing. Santilli et al. (2005) reported muscle activation time as a percentage of the stance phase normalised to peak muscle activity during the recorded trial. The different methodologies implemented have also led to differing findings. When investigating gluteus medius muscle onset times an earlier but not significantly different onset was found in the CAI group when compared to the healthy matched control (Feger et al., 2015). Koldenhoven et al. (2016) reported significantly increased gluteus medius RMS area under the curve during the 100 ms pre-HS and a higher sEMG amplitude during the final 50% stance and the first 25% of the swing phase curve in the CAI group when compared to the healthy control.

When investigating tibialis anterior and peroneus longus muscle sEMG, earlier muscle onset times were reported in the CAI group when compared to the healthy matched control. A longer peroneus longus activation duration was also observed in the CAI group across the entire stride cycle, however, no significant differences were observed in sEMG amplitudes (Feger et al., 2015). In contrast, Hopkins et al. (2012) observed increased tibialis anterior sEMG amplitudes following HS and during midstance, as well as increased peroneus longus amplitudes at HS and toe off in the ankle instability group when compared to healthy controls. Significantly increased peroneus longus and lower tibialis anterior RMS areas in the 100 ms prior to IC are also reported in CAI populations compared to healthy control groups (Koldenhoven et al., 2016). When investigating soleus activation during treadmill walking no significant differences were observed in muscle activity from 200 ms pre-HS to 200 ms post HS or in muscle latency at HS (Delahunt et al., 2006a).

Rectus femoris sEMG has been reported to display increased activity prior to HS though no differences following HS were documented (Delahunt et al., 2006a). In contrast, no significant differences were observed in sEMG amplitudes of the rectus femoris between groups though an earlier muscle onset time was reported in the instability group however this was not significant (Feger et al., 2015). This was also the case with the biceps femoris muscle.

Reporting of gastrocnemius sEMG differs in terms of location with Feger et al. (2015) reporting lateral gastrocnemius and Koldenhoven et al. (2016) reporting medial gastrocnemius. Earlier muscle onset time was reported in the lateral gastrocnemius of the instability group though this was not significant (Feger et al., 2015). Increased medial gastrocnemius RMS area in the 100 ms prior to IC was reported in the CAI group compared to the healthy control (Koldenhoven et al., 2016).

The differences in findings within muscles due to differing methodologies enables limited conclusions to be drawn by clinicians when implementing rehabilitation and prevention strategies, therefore, further investigation is needed across the entire gait cycle. Equally, no research to date has compared the muscle activation of the affected limb to the unaffected limb or investigated differences in the unaffected limb when compared to the healthy control group.

2.12.2 Kinematic Findings in Walking and CAI

Existing research into walking in individuals with CAI uses differing methodologies. Previous literature investigating CAI during walking has modelled the foot as one rigid segment (Monaghan et al., 2006; Stebbins et al., 2006). De Ridder et al. (2013) appears to be the first study to analyse walking using a multi-segmental foot model, comparing the use of the Ghent Foot Model to a rigid foot model in participants with CAI, copers (no symptoms of instability after a recent ankle sprain) and control participants. Results led the authors to conclude that the multi-segmental foot model provided greater details of the intricacies of the foot, showing differences between segments when comparing groups. Research by Monaghan et al. (2006) reports movement of the hip, knee and ankle with use of a single segment foot model, however, to the author's knowledge no research combines a full body marker set with a multi-segmental foot model. Similarly, to date, no research documents trunk kinematics during gait. This may provide further insight into kinematic differences present in those with instability.

Differing kinematic variables have been reported in the existing literature with some reporting displacement at various time points (Chinn et al., 2013; Delahunt et al., 2006a;

Drewes et al., 2009a; Wright et al., 2013b), whilst others document stride to stride variability within gait (Herb et al., 2014; Terada et al., 2015) and minimal research looking at angular velocity (Monaghan et al., 2006). Differing methodologies are implemented within these studies and as with sEMG, this produces different outcomes. Few studies compared differences in displacement of the hip and knee. Monaghan et al. (2006) reported no significant differences in hip and knee kinematics in any of the three planes of motion. Similar findings were observed when comparing the CAI group's affected limb and the control group's left limb when walking on a treadmill. Again, no significant differences were observed in hip or knee kinematics (Delahunt et al., 2006a), however, both studies only compared the CAI affected limb with the left leg of the control group - no comparison was made to the unaffected limb of the CAI group.

In sagittal plane motion an increased ankle plantarflexion has previously been reported from 42 to 51% of the gait cycle in individuals with CAI when compared to healthy controls (mean difference 2.9˚ ± 0.2˚) (Chinn et al., 2013). Conflicting research has observed no significant difference in sagittal plane rearfoot kinematics between groups (Wright et al., 2013b). Though these both use differing methodologies with Chinn et al. (2013) reporting shod treadmill walking and Wright et al. (2013b) reporting barefoot overground walking.

Increased ankle inversion in the CAI affected limb compared to the healthy control limb has been documented in multiple studies at 100 ms prior to HS to 200 ms post-HS (ankle inversion increased approximately 6-7˚) (Monaghan et al., 2006), throughout the gait cycle (rearfoot inversion 2.07° ± 0.29°) (Drewes et al., 2009a), at HS (forefoot inversion mean difference 2.86° SE = 0.93) (Wright et al., 2013b), (ankle inversion 2.10° vs −1.43°), 50 ms prior to (ankle inversion 1.69° vs −1.43°) and 50 ms post HS (ankle inversion –

0.09° vs −2.78°) (Delahunt et al., 2006a). In contrast, a greater ankle eversion was reported in the CAI group at 11-73% stance (average difference 2.17°) (De Ridder et al., 2013), and no significant differences were observed in rearfoot frontal plane motion (Wright et al., 2013b). Although the majority of these studies appear to be in agreement, many utilise a single segment foot model (Delahunt et al., 2006a; Drewes et al., 2009a; Monaghan et al., 2006), potentially making inadequate conclusions of movement of the intersegmental motion of the foot. This was confirmed by De Ridder et al. (2013) who compared the use of the Ghent multi-segmental foot model and a rigid foot model for comparing walking in individuals with CAI and healthy controls. They observed a more everted foot position with the rigid foot model but a more inverted position for the medial forefoot within the CAI group. Results led the authors to conclude that the multi-segmental foot model may provide greater details of the intricacies of the foot by showing differences between segments.

Significantly increased inversion velocity was also observed at 5 ms prior to and post HS and between 150-195 ms post-HS (Monaghan et al., 2006). No other research seems to investigate velocities during walking. These may provide greater information into injury-related variables.

De Ridder et al. (2013) used SPM to compare foot kinematics between participants with CAI, copers and controls, identifying exact time periods of significantly increased forefoot inversion in the CAI group compared to the control group from 87% to 98% of stance phase (average difference of 9.42˚) and significantly increased inversion in the coper group when compared to the control from 10% to 83% of the stance phase (average difference of 7.42˚). Prior research reports joint angles and muscle activation characteristics at discrete time points during walking (Koldenhoven et al., 2016; Monaghan et al., 2006), rather than whole kinematic time-series curves.

No comparison has been made to date between the affected and the unaffected limb of the CAI group to determine if a compensatory strategy is adopted or to compare the unaffected limb of the CAI group to a healthy matched control. Knowledge of this would provide increased information to clinicians on rehabilitation and injury prevention strategies. No research currently investigates trunk kinematics during walking in individuals with CAI.

2.12.3 Testing Protocols in Walking and CAI

Research investigating walking kinematics uses differing protocols with some literature observing walking barefoot (Monaghan et al., 2006; Wright et al., 2013b) whilst others observe shod (Chinn et al., 2013; Herb et al., 2014). Significantly different muscle activation strategies have been observed when comparing shod and barefoot walking. (Scott, Murley, & Wickham, 2012). A study comparing muscle activation during walking in a flexible sole running shoe, a stability running shoe and barefoot observed a significantly increased tibialis anterior EMG amplitude in both the flexible sole shoe (21% increase) and stability shoe (24% increase) when compared to barefoot walking (Scott et al., 2012). A decreased peroneus longus muscle activation was observed when shod (flexible sole – 20% decrease; stability sole- 16% decrease) compared to barefoot. Additionally, time to peak activation was earlier in both muscles within both shod conditions compared to barefoot (Scott et al., 2012). The use of shoes has also been found to impact ankle, knee and hip kinematics though agreement of changes is inconsistent with differing shoe types being adopted (Morio, Lake, Gueguen, Rao, & Baly, 2009; Scott et al., 2012). There appears to be no set guidelines for the prescription of footwear with differing footwear being proven to have differing effects on kinematics. Analysis of barefoot movement may prove a more valid representation of movement in individuals with CAI. Although ecological validity will reduce this may provide more useful information for subsequent intervention strategies.

Some studies analysed gait kinematics and muscle activation during walking over ground (De Ridder et al., 2013; Monaghan et al., 2006; Wright et al., 2013b) whilst others used treadmills for gait analysis (Chinn et al., 2013; Herb et al., 2014). Again, these methodological differences have been associated with differing outcomes with less dorsiflexor, knee extensor and hip extensor moments observed during treadmill walking when compared to walking over ground. Along with lower muscle activity of the tibialis anterior during stance, and lower hamstrings, vastus medialis and adductor longus during early and mid-swing phase and higher activity for terminal swing (S. J. Lee & Hidler, 2008). Results of walking over ground may prove a more valid representation of everyday walking in individuals with CAI.

2.13 Landing and Ankle Instability

A systematic review reported that during a typical basketball game males performed 41-56 jumps and females' 19-43 jumps (Taylor, Wright, Dischiavi, Townsend, & Marmon, 2017). Following a jump is the impact phase of landing within which the downwards momentum of the body must be reduced to zero (Lees, 1981). Jump and landing strategies have been reported in volleyball for offensive and defensive movements (Taylor et al., 2017). Offensively, 84% of jumps are performed from 2 feet whilst 55% of landings are two-footed. Defensively, 99% of jumps are 2 footed and 57% of landings are bilateral. Landing is commonly reported as a mechanism for lateral ankle sprains. Of the

39% of noncontact ankle sprains incurred in the English Premier League, landing was described as the mechanism of 36% of ankle sprains (Woods et al., 2003).

Landing can be split into three phases; the pre-landing (the period of time before IC), the impact absorption phase and the balance phase (Lees, 1981). The pre-landing phase of landing is thought to be feedforward motor control where individuals use pre-programmed mechanisms to anticipate joint loading and control their centre of gravity to maintain joint stability with the approaching ground contact (Delahunt, Monaghan, & Caulfield, 2007). Literature suggests that pre-IC muscle activity occurs at approximately 200 ms pre-ground contact (pre-landing) - this response is thought to be modulated by vision (Santello, 2005), therefore, this period of the movement may play a particular role in injury prevention. This phase is referred to as the reactive phase of landing (Doherty et al., 2014). Landing from a jump takes place over approximately 1 second, however, the impact absorption phase is known to last for 150-200 ms. Beyond this point downwards momentum is reduced and the rest of the action is concerned with the maintenance of balance (Lees, 1981). It has been summarised that an ankle sprain can occur as early as 40 ms after IC (Fong, Chan, Mok, Yung, & Chan, 2009b).

Some research to date has investigated the differences in landing kinematics (Brown et al., 2008; Brown, Bowser, & Simpson, 2012; Caulfield & Garrett, 2002; De Ridder, Willems, Vanrenterghem, Robinson, & Roosen, 2015a; Delahunt, Monaghan, & Caulfield, 2006b; Doherty et al., 2016c; Gribble & Robinson, 2009, 2010; Kipp & Palmieri-Smith, 2012; Wright, Arnold, & Ross, 2016), kinetics and GRF data (Brown et al., 2008; Brown et al., 2012; Caulfield & Garrett, 2002; De Ridder et al., 2015a; De Ridder et al., 2015b; Delahunt et al., 2006b, 2007; Doherty et al., 2014; Doherty et al., 2016c; Gribble & Robinson, 2009, 2010; Kipp & Palmieri-Smith, 2012; Kunugi, Masunari, Yoshida, & Miyakawa, 2017; Ross

& Guskiewicz, 2004; Wright et al., 2016) between individuals with and without ankle instability. Limited research has investigated the differences in muscle activation between groups during landing (Brown, Ross, Mynark, & Guskiewicz, 2004; Caulfield, Crammond, O'Sullivan, Reynolds, & Ward, 2004; Delahunt et al., 2006b, 2007; Kunugi et al., 2017). It is necessary to have a good understanding of the possible biomechanical differences that may exist between individuals with instability and healthy controls to better inform preventative and rehabilitation strategies.

2.13.1 Electromyographic Findings in Landing and CAI

Limited research to date has compared muscle activation during landing (Brown et al., 2004; Caulfield et al., 2004; Delahunt et al., 2006b, 2007; Kunugi et al., 2017). Many studies have investigated the tibialis anterior and the peroneus longus (Brown et al., 2004; Caulfield et al., 2004; Delahunt et al., 2006b, 2007; Kunugi et al., 2017), some studies have investigated the soleus (Brown et al., 2004; Caulfield et al., 2004; Delahunt et al., 2006b, 2007), some the lateral gastrocnemius (Brown et al., 2004; Kunugi et al., 2017). Delahunt et al. (2006b, 2007) appear to be the only authors to investigate the rectus femoris. Kunugi et al. (2017) also investigated the peroneus brevis and the medial gastrocnemius muscle activations. No studies appear to investigate any muscles more proximal than the rectus femoris. Given that proximal adaptations have been suggested to occur in individuals with ankle instability (Hubbard et al., 2007; Tropp et al., 1985) this may provide valuable information to guide preventative and rehabilitative strategies.

Different methods have been implemented in each of these studies with Brown et al (2004) investigating muscle onset times and muscle activation. Whilst other studies simply investigate muscle activation (Brown et al., 2004; Caulfield et al., 2004; Delahunt et al., 2006b, 2007; Kunugi et al., 2017). Normalisation strategies also differ between studies with Brown et al. (2004) and Kunugi et al. (2017) comparing to maximal voluntary contractions whilst others compare to peak activity muscle activation from the jump landing trials (Caulfield et al., 2004; Delahunt et al., 2006b, 2007).

Caulfield et al. (2004) observed no significant differences for soleus or tibialis anterior activity pre or post-IC, however peroneus longus activation was significantly reduced pre-IC. Similarly, Delahunt et al. (2006b) also observed a significant decrease in peroneus longus muscle activation pre-IC and no significant differences were observed post-IC or in the other muscles investigated (rectus femoris, tibialis anterior or soleus). This is contrary to future research by the same author (Delahunt et al., 2007) who observed significantly increased rectus femoris, tibialis anterior and soleus muscle activation prior to and following IC, but observed no significant differences in peroneus longus activation. The later study investigated a lateral hopping movement rather than a single leg drop which may explain the differences between studies. Kunugi et al. (2017) also observed a significantly decreased peroneus longus muscle activation, however, they reported this from 75 ms prior to IC to 60 ms post-IC. Alongside this, they also reported a significant reduction in peroneus brevis muscle activation from 151 ms pre-IC to 116 ms post. This is the only study to date to report the peroneus brevis muscle activation during landing. In contrast to other literature, a reduction in tibialis anterior muscle activity was observed from 69 ms to 203 ms following IC. Another study with different findings is that of Brown et al. (2004) who observed no significant differences in tibialis anterior, peroneus longus or soleus muscle activation prior to landing. Following landing, significantly increased soleus muscle activation was observed in the stable group when compared to the instability group during the 1000 ms post landing. Existing EMG data shows little agreement between studies. This may again be due to different time points under analysis and differences in single leg landing protocols.

2.13.2 Kinematic Findings in Landing and CAI

A number of studies document ankle kinematics using a rigid segment foot model (Brown et al., 2008; Brown et al., 2012; Caulfield & Garrett, 2002; De Ridder et al., 2015a; Delahunt et al., 2006b, 2007; Doherty et al., 2014; Gribble & Robinson, 2009, 2010; Kipp & Palmieri-Smith, 2012; Wright et al., 2016). When investigating rigid model ankle kinematics some authors have observed no significant differences between individuals with ankle instability and healthy controls (Brown et al., 2008; Brown et al., 2012; De Ridder et al., 2015a; Doherty et al., 2016c; Gribble & Robinson, 2009, 2010). In contrast, one study observed increased inversion from 200 ms to 95 ms prior to IC, less dorsiflexion from 90 ms to 200 ms post IC and decreased dorsiflexion velocity from 50 ms to 125 ms post IC (Delahunt et al., 2006b). A further study also observed increased inversion from 200 ms to 95 ms prior to IC (Delahunt et al., 2007) this was postulated to increase lateral ankle sprain risk in individuals with instability. Caulfield & Garrett (2002) observed increased ankle dorsiflexion from 10 ms pre-landing to 20 ms post-landing which was proposed as a learned adaptation to improve protection to the lateral ligament complex. Wright et al. (2016) used the Oxford Foot Model (a multi-segmental foot model) and also observed an increased hindfoot dorsiflexion position when compared to the healthy control group at IC. They, however, proposed that this position may in fact increase instability by decreasing the time available for the joint to absorb impact forces and suggested landing with increased plantarflexion allows increased range of motion for force attenuation to occur. The use of a multi-segmental foot model is thought to provide increased detail of intricacies between the joints of the foot (De Ridder et al., 2015b).

When investigating knee kinematics again several authors have observed no significant differences between groups (Brown et al., 2008; De Ridder et al., 2015a; Delahunt et al., 2006b, 2007; Doherty et al., 2016c). Caulfield & Garrett (2002) observed an increased knee flexion in the instability group when compared to controls from 20 ms pre-landing to 60 ms post-landing which was again suggested to be a learned adaptation. Comparatively, decreased knee flexion has also been observed pre-impact (Gribble & Robinson, 2010) and when impacting the ground (Gribble & Robinson, 2009) this will increase the height of the centre of mass from the ground and may be contributory towards increased instability.

Higher up the kinetic chain, analysis of the hip has observed less external rotation from 200 ms to 55 ms prior to IC, although authors were unsure as to the reason for this, it does confirm that proximal adaptations may exist in those with ankle instability (Delahunt et al., 2006b). Doherty et al. (2016c) observed increased hip flexion from 148 ms prior to IC to 4 ms following IC which was speculated as a potential method to reduce impact on contact. Further research is needed to combine EMG higher up the kinetic chain with kinematics to further investigate the proximal adaptations that may be present in individuals with ankle instability.

2.13.3 Testing Protocols in Landing and CAI

Several methodological differences exist between studies investigating single-leg landing which may help to explain the differences that exist between findings. Some studies have performed a single leg drop task off a set height box. However the height of the box changes between studies with 40 cm (Caulfield & Garrett, 2002; Caulfield et al., 2004; Caulfield & Garrett, 2004; De Ridder et al., 2015b; Doherty et al., 2014; Doherty et al., 2016c; Wright et al., 2016), 35 cm (Delahunt et al., 2006b), 32 cm (Brown et al., 2008) and 30 cm (Kunugi et al., 2017) all being used in the literature. No justification of selected height is given. Some methods differ between the instructed jump or step off the box.

Some studies ask the participant to stand with the test leg relaxed and non-weight bearing (Caulfield & Garrett, 2002; Caulfield et al., 2004; Caulfield & Garrett, 2004), whilst others perform jumps from test leg to test leg (Kunugi et al., 2017). Other studies first perform maximal jump heights and then ask participants to jump forward from 2 feet and land on one foot reaching 50% of jump height (Brown et al., 2004; Gribble & Robinson, 2010). Again there is some deviation with this method, with some authors specifying 40% of maximal jump height (De Ridder et al., 2015a) and others between 50 and 55% (Ross & Guskiewicz, 2004; Wikstrom, Tillman, Chmielewski, Cauraugh, & Borsa, 2007).

The duration of balance following the single leg land also differs between studies. With some asking participants to balance for just 2 seconds on landing (Kipp & Palmieri-Smith, 2012) and others asking participants to balance as long as 20 seconds following landing (Ross & Guskiewicz, 2004). Others do not specify the duration of balance following landing (Brown et al., 2004; Brown et al., 2012; Caulfield & Garrett, 2002, 2004; Delahunt et al., 2006b, 2007), reducing study repeatability and possibly impacting results.

Some studies asked participants to jump barefoot (Caulfield & Garrett, 2002; Caulfield et al., 2004; Caulfield & Garrett, 2004; De Ridder et al., 2015a; De Ridder et al., 2015b; Doherty et al., 2014; Doherty et al., 2016c; Wright et al., 2016), whilst others asked participants to jump shod however the shoes worn were not specified and it is unclear as to whether these were standardised between participants (Gribble & Robinson, 2010; Ross & Guskiewicz, 2004). Others do not specify whether jumps were performed barefoot or in shoes (Brown et al., 2004; Brown et al., 2008; Brown et al., 2012; Delahunt et al., 2006b, 2007; Gribble & Robinson, 2009; Wikstrom et al., 2007). Clear differences have been documented between kinematics, kinetics and muscle activation during landing shod or barefoot (Hong, Yoon, Kim, & Shin, 2014; Yeow, Lee, & Goh, 2011) so this must be reported to aid the reproduction of study results and make interpretation of findings clearer.

Some papers instruct participants on hand positioning with some instructing hands on hips during landing (De Ridder et al., 2015a; De Ridder et al., 2015b; Gribble & Robinson, 2009, 2010; Kunugi et al., 2017; Wikstrom et al., 2007). Although this may improve repeatability, this position lacks external validity, and very rarely will athletes adopt this position during landing in sport. These studies are summarised in Table 2.1.

Table 2.1 Summary of landing studies comparing those with ankle instability

Authors	Outcome variable	Protocol
Brown et al. (2004)	- EMG – Tibialis anterior, Peroneus longus, Lateral gastrocnemius, Soleus - Time to stabilisation	- Shoes/barefoot not specified - Maximum jump height recorded - Forward jump to 50% of the maximum jump height - 2 foot to 1-foot jump - Hold not specified
Brown et al, (2008)	- Force plate – GRF - Kinematics – Ankle and knee (sagittal and frontal plane)	- Shoes/barefoot not specified - Single leg drop jump from 32 cm box instructed not to jump ‘‘up” off the box but instead to ‘‘step off” - Approximately 3-second hold at the end of drop jump - No instructions were provided other than to make contact with the force plate
Brown et al, (2012)	- Kinematics - Ankle, knee and hip (sagittal, frontal and transverse plane)	- Shoes/barefoot not specified - Maximum jump height recorded - Single leg landings from a 50% maximum vertical jump in the anterior, lateral, and medial directions - 2 foot to 1-foot jump - Hold not specified
Caulfield & Garrett (2002)	- Kinematics - Ankle and knee (sagittal plane)	- Barefoot - Single leg jump from 40 cm box - Test leg relaxed and non-weight bearing – use contralateral limb to propel from box and land on test limb - Duration of hold and arm position not specified
Caulfield & Garrett (2004)	- Force plate - GRF	- Barefoot - Single leg drop jump from 40 cm box - Test leg relaxed and non-weight bearing – use contralateral limb to propel from box and land on test limb - Duration of hold and arm position not specified
Caulfield et al. (2004)	- EMG – Tibialis anterior, Peroneus longus, Soleus	- Barefoot - Single leg drop jump from 40 cm box and jump for distance - SL Box drop -Test leg relaxed and non-weight bearing – use contralateral limb to propel from box and land on test limb (free to select own landing technique) - Jump for distance - Test leg relaxed used opposite limb to propel themselves forwards to land on the test leg (self-selected their landing technique) - 3-second duration of hold
De Ridder et al. (2015a)	- Kinematics - Ankle, knee and hip (sagittal, frontal and transverse plane) - SPM analysis	- Barefoot - Forward jump - 2 footed forward jump to 40% of subjects height - jumping over a 30 cm hurdle - land on test leg - Lateral jump – 2 footed lateral jump to 33% of subjects height - jumping over a 15 cm hurdle - land on test leg - Hands free in flight but on hips in landing - Maintain balance for 5 seconds

De Ridder et al. (2015b)	- Kinematics – rigid foot vs Ghent foot model - Force plate – GRF - SPM analysis	- Barefoot - Single leg drop jump from 40 cm box and maximal sideward jump - Single leg drop jump – starting on opposite leg – step down onto test limb - Maximal sideward jump – starting on opposite foot – max sideward jump landing on test limb - Maintain balance for 3 seconds - Hands on hips throughout trial
Delahunt et al. (2006b)	- EMG – Tibialis anterior, Peroneus longus, Soleus, Rectus femoris - Kinematics - Ankle, knee and hip angular displacements and velocities - Force plate – GRF	- Shoes/barefoot not specified - Single leg drop jump from 35 cm box - Test leg relaxed and non-weight bearing - Test leg relaxed and non-weight bearing – use contralateral limb to propel from box and land on test limb - Duration of hold and arm position not specified
Delahunt et al. (2007)	- EMG – Tibialis anterior, Peroneus longus, Soleus, Rectus femoris - Kinematics - Ankle, knee and hip angular displacements and velocities - Force plate – GRF	- Shoes/barefoot not specified - Lateral hop – starting 30cm from force plate hop laterally onto and medially off the centre of the force plate at a self-selected velocity - Duration of hold and arm position not specified
Doherty et al. (2014)	- Kinematics - Ankle, knee and hip (sagittal, frontal and transverse plane) - Force plate – GRF	- Barefoot - Single leg drop jump from 40 cm box - Test leg non-weight bearing – drop forward onto test leg (free to select own landing technique) - Maintain balance for 4-6 seconds
Doherty et al. (2016c)	- Kinematics - Ankle, knee and hip (sagittal, frontal and transverse plane) - Kinetics – sagittal plane ankle, knee and hip moments and joint stiffness - Force plate – GRF	- Barefoot - Single leg drop jump from 40 cm box - Test leg non-weight bearing – drop forward onto test leg (free to select own landing technique) - Maintain balance for 4-6 seconds
Gribble & Robinson (2009)	- Kinematics - Ankle, knee and hip (sagittal plane) - Time to stabilisation	- Shoes/barefoot not specified - Maximum jump height recorded - Forward jump to 50% of the maximum jump height - 2 foot to 1-foot jump - Assumed hands on hips position on landing and maintained balance for 5 seconds
Gribble & Robinson (2010)	- Kinematics - Ankle, knee and hip (sagittal plane) - Time to stabilisation	- Shoes make/model not specified - Maximum jump height recorded - Forward jump to 50% of the maximum jump height - 2 foot to 1-foot jump - Assumed hands on hips position on landing and maintained balance for 5 seconds

Kipp & Palmieri-Smith (2012)	- Kinematics - Ankle, knee and hip (sagittal plane) - Kinetics- sagittal plane ankle, knee and hip moments and joint stiffness	- Shoes/barefoot not specified - Participants initiated a forward jump off 2 feet over a 15 cm box to land on a single leg. Distance was normalised to leg length from greater trochanter to lateral malleolus - Maintain balance for 2 seconds
Kunugi et al. (2017)	- EMG – Tibialis anterior, Peroneus longus, Peroneus brevis, Medial and Lateral gastrocnemius - Force plate – GRF	- Barefoot - Diagonal single leg drop jump from 30 cm box - Test leg to test leg - Balanced for 20 seconds with hands on their hips
Ross & Guskiewicz (2004)	- Force plate – GRF, sway, time to stabilisation	- Shoes make/model not specified - Maximum jump height recorded - Forward jump to between 50% and 55% of the maximum jump height - 2 foot to 1-foot jump - Balanced for 20 seconds - Did not control for arm position, trunk flexion, or lower extremity flexion during
Wikstrom et al. (2007)	- Force plate – GRF	- Shoes/barefoot not specified - Maximum jump height recorded - Forward jump to between 50% and 55% of the maximum jump height - 2 foot to 1-foot jump - Balanced for 10 seconds - Hands on hips
Wright et al. (2016)	- Kinematics – Forefoot and hindfoot (frontal and sagittal plane) - Force plate – GRF and time to stabilisation	- Barefoot - Single leg drop jump from 40 cm box - Balanced for 10 seconds - Arm position not specified

2.13.4 Accidental Sprains During Laboratory Testing - Landing

Few studies have analysed ankle sprains or episodes of giving way that have occurred during laboratory testing in order to gain a greater understanding of the mechanism of lateral ankle sprains (Y. Li, Ko, Zhang, Brown, & Simpson, 2018; Terada & Gribble, 2015). Y. Li et al. (2018) reported two ankle sprains in young female participants performing a jump landing onto a 25˚ laterally tilted force platform. Recording 3D kinematics, kinetics and muscle activity of the lower extremity. No pain or injury was reported following these episodes of giving way. They reported increased ankle inversion (13°–17° vs 10°–12°), increased ankle internal rotation (~10° vs 2°–7°) and less hip abduction (3°–5° vs 6°–7°) pre-landing and at IC when compared to the previous successful trials. In participant 2 they observed increased hip abduction (9˚ vs 6˚), increased peak ankle inversion velocity (927°/s vs. 528°/s) and delayed peroneus longus activation prior to landing. Comparatively, Terada and Gribble (2015) reported an accidental lateral ankle sprain during a bilateral stop-jump task. The participant suffered a mild lateral ankle sprain in the 3rd of 5 trials the participant reported that this mimicked the usual recurrent ankle sprains that they experienced as a CAI sufferer. Prior to injury a 33% greater knee adduction, 22% greater hip abduction, 11% less ankle plantarflexion and 43% less peak knee flexion were observed when compared to the previous 2 non-injury trials. The COM was also 2 cm higher and 2 cm shifted towards the non-injured side. During the injury period they observed 35% greater peak ankle inversion, 30% higher peak knee adduction, 22% higher peak hip abduction, 19% lower peak knee flexion and 25% lower peak hip flexion. Along with a 7 cm higher and 3 cm lateral shift towards the non-injured side in COM location. Both studies report significant differences in joint angular displacements and angular velocities in distal but also in proximal joint kinematics suggesting differences in both may result in episodes of giving way or contribute to recurrent ankle sprains.

2.14 Cutting and Ankle Instability

Cutting manoeuvres are often used during sport to evade markers and react to opposition's movement (Bloomfield, Polman, & O'Donoghue, 2007). An audit of injuries in football in the English Premier League found 39% of ankle sprains to occur in non-contact situations. Twisting and turning accounted for 21% of these injuries (Woods et al., 2003). Changing direction involves braking in the original direction (forward) followed by translation and reorientation into the new direction (Havens & Sigward, 2015). For this to occur ground reaction forces and ground reaction force impulse and position of the centre of mass must be positioned posteriorly to the centre of pressure and similar adjustments made in the medial-lateral direction to move away from the original direction (Havens & Sigward, 2015). When observing movements of 55 Premier League players during a match, a total of 727 ± 203 turns were reported, with midfielders performing fewer than defenders and strikers (Bloomfield et al., 2007). These turns were further broken down into 0-90˚, 90-180˚, 180-270˚ and 270-360˚. The most common category was 0-90˚ (left side 303.2 ± 99.3; right side 305.8 ± 104.7) followed by 90-180˚ (left side 49.3 ± 20.1; 45.2 ± 19.4)(Bloomfield et al., 2007).

Limited research has investigated cutting mechanics in individuals with ankle instability. Where research has been conducted, differences in protocols and variables under investigation exist. Some studies have investigated electromyography variables (Fuerst, Gollhofer, Lohrer, & Gehring, 2018; Koshino et al., 2016; Son, Kim, Seeley, & Hopkins, 2017; Suda & Sacco, 2011), some kinematics (Fuerst et al., 2018; Koshino et al., 2014; Koshino et al., 2016; Son et al., 2017), some joint kinetics and GRF data (Dayakidis &

Boudolos, 2006; Fuerst et al., 2018; Koshino et al., 2014; Koshino et al., 2016; Son et al., 2017; Suda & Sacco, 2011) and one foot pressures (Huang, Lin, Kuo, & Liao, 2011).

2.14.1 Electromyographic Findings in Cutting and CAI

When investigating electromyography several muscles have been investigated. These include the tibialis anterior (Fuerst et al., 2018; Koshino et al., 2016; Son et al., 2017; Suda & Sacco, 2011), peroneus longus (Fuerst et al., 2018; Koshino et al., 2016; Son et al., 2017; Suda & Sacco, 2011), lateral gastrocnemius (Fuerst et al., 2018; Suda & Sacco, 2011), medial gastrocnemius (Koshino et al., 2016; Son et al., 2017), gluteus medius and maximus (Koshino et al., 2016; Son et al., 2017), soleus (Fuerst et al., 2018), rectus femoris (Koshino et al., 2016), vastus lateralis (Son et al., 2017), medial hamstring (Son et al., 2017) and semitendinosus (Koshino et al., 2016). Suda et al. (2011) reported muscle onset times. They reported differences in motor strategies with the control group activating the lateral gastrocnemius significantly earlier followed by the peroneus longus and tibialis anterior whilst the instability group activated the lateral gastrocnemius and the peroneus longus at the same time followed by the tibialis anterior significantly later this was attributed to greater individual variance in the instability group. When comparing muscle activation a number of different methods are used to normalise data with Suda & Sacco (2011) and Koshino et al. (2016) using a maximal voluntary isometric contraction, Son et al. (2017) normalising to the mean three seconds of an isometric double legged squat and Fuerst et al. (2018) normalising to mean RMS value of five stride cycles during straight line running. During a side turn movement Koshino et al. (2016) observed no significant differences in mean EMG activity between groups, however, during the side cutting movement observed increased mean activity of the medial gastrocnemius in the instability group when compared to the control group during 10-

30% stance. Fuerst et al. (2018) observed no significant differences in muscle activation between groups for the 45-degree cut, during the 180-degree turn they observed significantly higher values for peroneus longus activation prior to IC in the FAI group when compared to the control group. No other significant differences were observed for muscle activation. Comparatively, Suda & Sacco (2011) observed lower peroneus longus activation in the instability group in the 50 ms prior to IC. Lastly, Son et al. (2017) observed several significant differences during the stance phase of the cutting movement. Decreased tibialis anterior activation was observed at 36-100%, decreased peroneus longus activation at 0-66%, decreased medial gastrocnemius activation at 23-65%, increased vastus lateralis activation at 2-21% and decreased activation during 44-60%, increased gluteus medius activation during 3-14% and decreased activation during 35-45% and lastly decreased gluteus maximus activation during 24-71%.

2.14.2 Kinematic Findings in Cutting and CAI

Research investigating kinematics during change of direction manoeuvres has used a rigid foot model and reported movement of the ankle, knee and hip (Koshino et al., 2014; Koshino et al., 2016; Son et al., 2017). Fuerst et al. (2018) used hip and knee markers however did not document movement above the ankle. They reported maximum inversion angles between groups but observed no significant differences between groups ($p = 0.059$) though did report post hoc comparisons between groups. In contrast, Son et al. (2017) observed several differences in ankle kinematics; less plantarflexion was observed at 0-24% and 83-100% stance and less dorsiflexion was observed during 34-69% stance. They also observed less inversion during 6-38% stance. These differences were suggested to place increased stress on the tibiotalar articular cartilage and result in a loss of mechanical advantage. Koshino et al. (2016) reported increased ankle inversion from pre-IC 200 ms to pre-IC 165 ms and from 78-100% stance which they suggested may be a predisposing factor to lateral ankle sprains. Previous research comparing a multi-segmental foot model to a rigid foot model has suggested that these findings may not be representative of the different segments of the foot (De Ridder et al., 2013; De Ridder et al., 2015b).

Several differences have been noted between individuals with ankle instability during cutting in the knee and hip. Koshino et al. (2016) observed increased hip flexion during the cross-cutting motion from 11-18% stance which was suggested as a means of lowering the centre of mass to improve dynamic stability. Son et al. (2017) also observed increased hip flexion during 3-100% stance in the instability group, along with increased knee flexion (5-36% and 72-88% stance), increased knee abduction (0-7%, 18-42% and 84-97% stance) and less hip abduction (10-20% stance). They suggested that the decreased hip abduction was an attempt to adopt a more vertical femoral position to maintain a close distance between the centre of mass and the centre of pressure. They also proposed that the increased hip and knee flexion may be an attempt to help attenuate the impact forces and disperse away from the 'unstable' ankle joint. Similarly, Koshino et al. (2014) also observed increased hip flexion from 6%-50% stance phase, increased hip abduction from the pre-IC 200 ms to 45% stance phase and increased knee flexion from 35%-64% and 69%-87% stance in the instability group. These findings suggest more proximal adaptations are made to either improve stability or help to distribute impact forces. However, no research seems to document movement superior to the hip. The kinetic chain principle would suggest that the trunk may also experience some proximal adaptations during movement.

2.14.3 Testing Protocols in Cutting and CAI

The movement under investigation differs in each study. A few studies have investigated a lateral shuffle movement whereby participants performed 2-3 shuffles one way before hitting the force plate and moving back in the opposite direction (Dayakidis & Boudolos, 2006; Huang et al., 2011; Suda & Sacco, 2011). Koshino et al. (2014; 2016) investigated a cross turn and crosscut movement. The cross-turn movement required participants to walk straight ahead at their natural walking speed, they then planted their foot on the force plate and changed direction to the side of the supporting leg at a 45-degree angle before proceeding 2.5 m. The crosscut movement required participants to jump forward onto the force plate (on hearing an audio cue) and land on the test limb they then performed a 45-degree crossover cut and proceeded to run for 2.5 m (Figure 2.11). Dayakidis & Boudolos (2006) investigated a v-cut movement (7 m forward run at a controlled approach speed of 5.0 ± 0.2 m/s, planting either left or right foot and cutting at a 45-degree angle). Within a study by Son et al. (2017) participants were asked to jump as high as they could and land on the force plate with their test leg only and side cut 90° to the contralateral side as quickly as possible. They were then asked to perform a 2 footed maximal vertical forward jump before performing a 90 degree cut to the contralateral side. Fuerst et al. (2018) controlled the straight line approach speed at a velocity of 4 ± 0.3 m/s and asked participants to perform a 45 degree, a 25 degree and a 180 degree cut. Differences in the cutting protocol from a straight line or jump approach and differences in cutting angle and technique may help to explain the differences in findings evident between studies. These studies are summarised in Table 2.2.

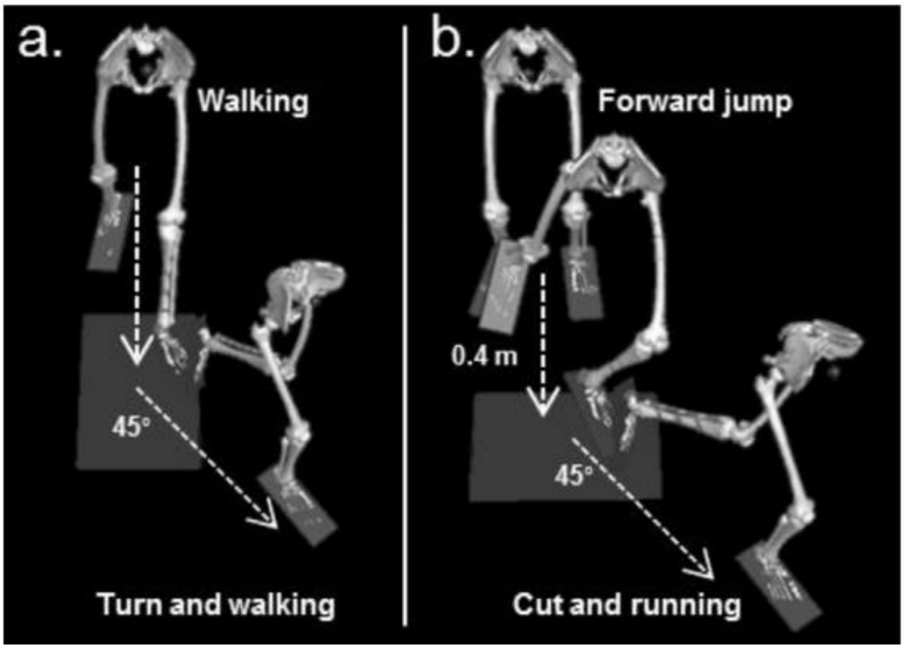


Figure 2.11 Cutting manoeuvresas performed in Koshino et al. (2014)

Table 2.2 Summary of cutting studies comparing those with ankle instability

Authors	Outcome variable	Protocol
Dayakidis & Boudolos (2006)	- Force plate- GRF	- Barefoot/ shoes not defined - V-cut - 45 ° v-cut movement (7 m forward run at a controlled approach speed of 5.0 ± 0.2 m/s, planting either left or right foot and cutting at a 45-degree angle) - Lateral shuffle – starting in a crouched position shuffle one side twice then immediately change direction back along the same line
Fuerst et al. (2018)	- Kinematics - Ankle (frontal plane) - Kinetics – Ankle moments (frontal plane) - EMG – tibialis anterior, soleus, Lateral gastrocnemius, Peroneus Longus	- Shoes (Adidas Spezial, ADIDAS AG, Herzogenaurach, Germany) - Straight approach run (velocity of 4 ± 0.3 m/s) followed by a 45 ° sidestep-cutting movement - Straight approach run (velocity of 4 ± 0.3 m/s) followed by a 25 ° crossover-cutting movement Straight approach run (velocity of 4 ± 0.3 m/s) followed by a 180 ° turning movement
Huang et al. (2011)	- Foot pressures	- Shoes (JUMP, Lu-Tung Corporation, Taipei, Taiwan) - Subjects were asked to run with a comfortable speed and perform the lateral shuffling as fast as possible
Koshino et al. (2016)	- Kinematics - Hip flexion, adduction, and internal rotation, knee flexion, and ankle dorsiflexion and inversion angles - EMG - Gluteus maximum, Gluteus medius, Rectus femoris, Semitendinosus, Peroneus longus, Tibialis anterior, Medial gastrocnemius - Force plate- GRF	- Shoes (Artic Mesh M, Adidas, Herzogenaurach, Germany) - Cross turn - Participants instructed to walk straight ahead at their natural walking speed – they then planted foot on the force plate and changed direction to the side of the supporting leg at a 45 ° angle before proceeding 2.5 m. - Crosscut - participants instructed to jump forward onto the force plate (on hearing an audio cue) and land on their test limb they then performed a 45 ° crossover cut and proceeded to run for 2.5 m
Koshino et al. (2014)	- Kinematics - Hip flexion, adduction, and internal rotation, knee flexion, and ankle dorsiflexion and inversion angles	- Shoes (Artic Mesh M, Adidas, Herzogenaurach, Germany) - Cross turn - Participants instructed to walk straight ahead at their natural walking speed – they then planted foot on the force plate and changed direction to the side of the supporting leg at a 45 ° angle before proceeding 2.5 m. - Crosscut - participants instructed to jump forward onto the force plate (on hearing an audio cue) and land on their test limb they then performed a 45 ° crossover cut and proceeded to run for 2.5 m
Son et al. (2017)	- Kinematics - Ankle, knee and hip (sagittal, frontal and transverse plane) - Force plate- GRF - EMG – Tibialis anterior, Peroneus longus, Medial gastrocnemius, Vastus lateralis, Medial hamstring, Gluteus maximum, Gluteus medius	- Barefoot/ shoes not defined - Participants were asked to jump as high as they could and land on the force plate with their test leg only and side cut 90˚ to the contralateral side as quickly as possible. They were then asked to perform a 2 footed maximal vertical forward jump before performing a 90 degree cut to the contralateral side.
Suda & Sacco, (2011)	- Force plate- GRF - EMG – Tibialis anterior, Peroneus longus, Lateral gastrocnemius	- Own volleyball shoes worn – not specified - Lateral shuffle – subjects shuffled to the side twice, hitting the force platform and returning the other direction - Subjects were instructed to perform the movement as quickly as possible

2.15 Determining Initial Contact Using Kinematics

When the use of a force plate is not suitable for the test movement or where unavailable for testing, several methods have been proposed to determine the point of IC and toe-off. Within this thesis access to a force plate is not possible therefore it will be necessary to use a rule to ensure a consistent and repeatable approach to be implemented. These use a range of variables for example, angular displacements, marker co-ordinates, marker displacements, angular velocities and angular accelerations (as detailed in Table 2.3).

Table 2.3 Summary of methods for determining foot strike and toe off using kinematic data (as defined in the literature)

Method	Foot strike	Toe off	Reference
Peak knee extension	First peak in knee extension	Second peak in knee extension	Dingwell, Cusumano, Cavanagh, and Sternad (2001); Fellin, Rose, Royer, and Davis (2010)
Foot vertical position	Minimum vertical position of the distal heel	Minimum vertical position of the 2nd metatarsal head	Fellin et al. (2010)
Foot-sacrum displacement	Maximum positive displacement in the direction of progression between the sacrum and distal heel	Maximum negative displacement, along anterior-posterior axis, between the 2nd metatarsal head and sacrum	Fellin et al. (2010)
Foot vertical velocity	Change in vertical velocity from negative to positive of the distal heel	Change in vertical velocity from negative to positive of the 2nd metatarsal head	Fellin et al. (2010)
Angular acceleration	Time of the local minimum of foot angular acceleration in the sagittal plane	Time of the local minimum of the shank angular acceleration in the sagittal plane	Fellin et al. (2010); Hreljac and Stergiou (2000)
Coordinate-Based Treadmill Algorithm	Maximum of $X_{heel} - X_{sacrum}$	Minimum of $X_{toe} - X_{sacrum}$	De Witt (2010); Zeni, Richards, and Higginson (2008)

Velocity based treadmill algorithm	Point when X component of the velocity vector for the heel marker changes from positive to negative	Point when X component of the velocity vector for the toe or heel markers changes from negative to positive	De Witt (2010)
Vertical component of toe marker acceleration and jerk	Local maximum in vertical toe marker acceleration (between previous HS and the next maximum local vertical heel marker position)	Toe marker jerk equal to zero using equation for linear interpolation: Time of event = t_1 + (J(t_1)/(J(t_1)-(J(t_2))*t_{int} t_1 = time of the sample with the last positive vertical jerk value t_2 = time at which the first negative jerk occurred t_{int} = time between samples (1/60 Hz), J(t_1) and J(t_2) = jerk values at t_1 and t_2.	De Witt (2010)
Foot velocity algorithm	A new signal, representing the foot centre, is created by calculating the midpoint of the heel and toe marker locations. Minimum of virtual marker vertical velocity	Peak of virtual marker vertical velocity	C. M. O'Connor, Thorpe, O'Malley, and Vaughan (2007)
Foot marker kinematics	The second of the W shaped minima of the foot vertical velocity curve in the Z (vertical) axis	The minimum position of the toe-markers in the Z axis	Mickelborough, van der Linden, Richards, and Ennos (2000); Sinclair, Edmundson, Brooks, and Hobbs (2011)

Malleolus position	Minimum position of the lateral malleolus in the Z axis	The minimum position of the toe-markers in the Z axis (as in Mickelborough et al. (2000).	Alton, Baldey, Caplan, and Morrissey (1998); Sinclair et al. (2011)
Vertical velocity and displacement of the foot markers	Time of the downward spike of the vertical velocity of the 1st metatarsal and the plateau in the displacement of the lateral malleoli marker in the Z axes	Onset of the rise in vertical displacement and velocity of the 1st metatarsal marker.	Schache et al. (2001); Sinclair et al. (2011)

A number of studies have compared the different methodologies. Fellin et al. (2010) compared the use of peak knee extension, foot vertical position, foot-sacrum displacement, foot velocity and angular acceleration to determine HS and toe off during overground and treadmill running. They concluded that foot vertical velocity was the most accurate method for determining FS in overground running followed by foot vertical position. In treadmill running however the foot vertical position was found to be most accurate followed by the foot vertical velocity. Both tasks found peak knee extension to be the best algorithm for determining toe off (Fellin et al., 2010). De Witt (2010) compared the use of the coordinate based treadmill algorithm and the velocity-based treadmill algorithm as proposed by Zeni et al. (2008) to vertical toe position (using acceleration and jerk). They concluded that the vertical toe position was the most accurate method (De Witt, 2010). C. M. O'Connor et al. (2007) compared the foot velocity algorithm (FVA) to the method proposed by Hreljac and Stergiou (2000) and observed more accurate results with the FVA method when compared to force plate readings and also noted increased ease of application. Following from these, one study compared the methods proposed by Mickelborough et al. (2000), C. M. O'Connor et al. (2007), Alton et al. (1998), Hreljac and Stergiou (2000), Zeni et al. (2008) and Dingwell et al. (2001) (detailed in Table 2.3) but also incorporated a force plate in order to validate their use against the recognised gold standard (Sinclair et al., 2011). They observed significantly lower average and absolute errors in the methods of Alton et al. (1998), C. M. O'Connor et al. (2007) and Dingwell et al. (2001) for HS and in the methods of Dingwell et al. (2001) at toe-off (Sinclair et al., 2011).

Several different methods have been proposed in the literature. Differing tasks of observation seem to require different methods for determining IC, therefore, this suggests that algorithms for determining IC may be task specific which may explain why the literature proposes no clear recommendations for this. Although the gold standard for determining IC remains the force platform, kinematics have been found to be a repeatable method when force plates are not available or appropriate for the nature of testing (Fellin et al., 2010; Handsaker, Forrester, Folland, Black, & Allen, 2016; Sinclair et al., 2011).

2.16 Rationale

Several rehabilitation and preventative strategies have been suggested in the literature for the reduction in incidence of lateral ankle sprains particularly in individuals with CAI. However, despite these strategies, the rate of ankle sprains in individuals with ankle instability remains high, resulting in high-cost implications and impact upon quality of life. Increasing knowledge of biomechanical quantities is thought to be extremely important in the development of injury prevention strategies and protective equipment. The kinetic chain principle states movement at one joint impacts the movement of another. This is reinforced by ankle taping and bracing studies that have observed proximal adaptations to an intervention (Cordova et al., 2010; DiStefano et al., 2008; Santos et al., 2004; Stoffel et al., 2010). To produce the most effective intervention strategy possible a full knowledge of kinematic movement patterns not just of the ankle joint is essential. This thesis will analyse three key movements in ankle instability. Walking, a task commonly associated with ankle sprain in the general population and landing and cutting which account for the two most common mechanism of ankle sprains in sporting situations. It is suggested that increased knowledge of possible differences in full body kinematics and muscle activation patterns during movements particularly prone to injury may highlight key areas for future intervention and prevention strategies but also potentially be a basis to identify individuals who may be at risk of future recurrent issues following an initial ankle sprain.

2.17 Aims

The main aims of this thesis were:

- To explore full time series whole body kinematics during walking, single leg landing and cutting
- To explore full time series muscle activation patterns during walking, single legged landing and cutting
- To explore discrete kinematic variables during walking, single leg landing and cutting
- To identify the relationship between observed significant differences and score on the identification of functional ankle instability questionnaire

2.18 Objectives

The main objectives of this thesis were:

- To measure whole body kinematics in individuals with chronic ankle instability and healthy control participants using three-dimensional motion analysis and exploring using statistical parametric mapping and discrete statistical methods
- To measure muscle activation patterns in individuals with chronic ankle instability and healthy control participants using surface electromyography and statistical parametric mapping
- To explore the relationship between significant differences and score on the Identification of Functional Ankle Instability questionnaire using regression analysis

2.19 Hypotheses

Due to the exploratory nature of this study general study hypotheses will be tested rather than individual experimental hypotheses (Table 2.4)

Table 2.4 Experimental and null hypotheses

Experimental hypotheses	Null hypotheses
H_1 - CAI participants will display modified kinematic movement patterns (SPM) during walking	**$H0_1$** - CAI participants will display no modification in kinematic movement patterns (SPM) during walking
H_2 - CAI participants will display modified muscle activation patterns (SPM) during walking	**$H0_2$** - CAI participants will display no modification in muscle activation patterns (SPM) during walking
H_3 - CAI participants will display modified discrete kinematic variables during walking	**$H0_3$** - CAI participants will display no modification in discrete kinematic variables during walking
H_4 - Significant differences observed during walking will be able to predict IdFAI questionnaire score	**$H0_4$** - Significant differences observed during walking will not be able to predict IdFAI questionnaire score
H_5 - CAI participants will display modified kinematic movement patterns (SPM) during single-leg landing	**$H0_5$** - CAI participants will display no modification in kinematic movement patterns (SPM) during single-leg landing
H_6 - CAI participants will display modified muscle activation patterns (SPM) during single-leg landing	**$H0_6$** - CAI participants will display no modification in muscle activation patterns (SPM) during single-leg landing

H_7 - CAI participants will display modified discrete kinematic variables during single-leg landing	**$H0_7$** - CAI participants will display no modification in discrete kinematic variables during single-leg landing
H_8 - Significant differences observed during single-leg landing will be able to predict IdFAI questionnaire score	**$H0_8$** - Significant differences observed during single-leg landing will not be able to predict IdFAI questionnaire score
H_9 - CAI participants will display modified kinematic movement patterns (SPM) during cutting	**$H0_9$** - CAI participants will display no modification in kinematic movement patterns (SPM) during cutting
H_{10} - CAI participants will display modified muscle activation patterns (SPM) during cutting	**$H0_{10}$** - CAI participants will display no modification in muscle activation patterns (SPM) during cutting
H_{11} - CAI participants will display modified discrete kinematic variables during cutting	**$H0_{11}$** - CAI participants will display no modification in discrete kinematic variables during cutting
H_{12} - Significant differences observed during cutting will be able to predict IdFAI questionnaire score	**$H0_{12}$** - Significant differences observed during cutting will not be able to predict IdFAI questionnaire score

Chapter 3.0 General Methods

Throughout this thesis, several core protocols will be used. These will refer back to the methods within this chapter.

3.1 Participants

Eighteen (14 males, 4 females) healthy controls (age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 (13 males, 5 females) participants with CAI (age: $\bar{\chi}$ = 22.0 ± 2.7 years, height: $\bar{\chi}$ = 176.8 ± 7.9 cm, mass: $\bar{\chi}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were used for Chapter 5, 6 and 7. Participants were included in the study if they were aged 18-35 and took part in team sport a minimum of twice a week (minimum of 30 minutes per session). Ethical approval was granted by the university ethics committee prior to testing. Written informed consent was obtained from participants and a health screen questionnaire was completed prior to participation.

Participants were recruited through convenience sampling. Participants were allocated into the control group or the CAI group based on results of the IdFAI questionnaire, where a score of ≥11 indicated ankle instability in accordance with IAC guidelines (Gribble et al., 2013). All participants completed the IdFAI questionnaire on a Google forms questionnaire. Anonymised data from this questionnaire was viewed by the investigator prior to data analysis to ensure group sizes were balanced where this was not the case additional participants were recruited. All participant data were anonymized so to avoid bias during the analysis phase. Inclusion criteria are outlined in Table 3.1. In the instance of bilateral ankle sprains, the involved limb was selected based on the participant's perception of greater instability. As the researcher was blinded to the questionnaire outcome, the affected limb could not be identified exclusively as either the dominant or non-dominant limb. Therefore, the control group were matched for dominance to the instability group (if the dominant limb is the affected limb the matched control will also be the dominant limb) to ensure that the dominance effect is accounted for within the analysis. Limb dominance was determined by asking which leg they would use to kick a ball (Hopkins et al., 2012). Mean IdFAI score for the control group was 3.71 ± 3.13 and 19.1 ± 6.25 in the CAI group's affected limb.

Table 3.1 Inclusion and exclusion criteria as outlined by the International Ankle Consortium (Gribble et al., 2013)

Inclusion	Exclusion
1. A history of at least 1 significant ankle sprain • Initial sprain at least 12 months prior • Initial sprain associated with inflammatory symptoms (pain, swelling, etc.) • Initial sprain resulted in at least 1 interrupted day of physical activity • Most recent injury must have occurred more than 3 months prior to study enrolment 2. History of the previously injured ankle joint "giving way," and/or recurrent sprain, and/or "feelings of instability." • Participants should report at least 2 episodes of giving way in the 6 months prior to study enrolment • Self-reported ankle instability confirmed with the use of the Identification of Functional Ankle Instability: score of ≥11 indicates ankle instability	1. History of previous surgeries to the musculoskeletal structures (i.e., bones, joint structures, nerves) in either lower extremity 2. History of fracture in either lower extremity requiring realignment 3. Acute injury to musculoskeletal structures of other joints of the lower extremity in the previous 3 months that impacted joint integrity and function (i.e., sprains, fractures), resulting in at least 1 interrupted day of desired physical activity 4. Regular use of orthotics

3.2 Identification of Functional Ankle Instability Questionnaire

The Identification of Functional Ankle Instability questionnaire (Simon et al., 2012) was used to confirm self-reported ankle instability. This can be viewed along with the questionnaire scoring in (Appendix D, Figure D.1). The questionnaire was transcribed onto computer-based questionnaire software to enable blinding of the researcher to scoring.

3.3 Height and Mass

Participants' anthropometric measurements of height and mass were recorded in accordance with the British Association of Sport and Exercise Science guidelines (Winter, Jones, Davison, Bromley, & Mercer, 2006). Height was measured to 0.1cm (Seca stadiometer 225b, Hamburg, Germany). Participants were required to stand barefoot with their feet together and their head in the Frankfort plane. The measurement was taken with inspiration where the headboard was brought down to compress the hair. Mass was measured to 0.1kg (Seca Electrical Column Digital Scales 780, Hamburg, Germany) and was taken with the participant barefoot and wearing light clothing.

3.4 Electromyography

Electromyographic data reported throughout the thesis were recorded bilaterally using a DataLINK data acquisition system (Biometrics Bluetooth unit W4X8, Biometrics Ltd, Gwent, UK) sampling at 1000Hz with pre-amplified electrodes (Biometrics Ltd, SX230-1000, gain x1000, bandwidth 20-450 Hz, noise < 5 μV, input impedance > 1015 Ω). Participants' skin was prepared for electrode placement and electrodes placed in accordance with SENIAM guidelines (SENIAM, 2004). Tibialis anterior electrodes were placed at a third of the line between the tip of the head of the fibula and the tip of the medial malleolus. Gluteus medius electrodes were placed halfway between the crista iliaca and the greater trochanter. The peroneus longus electrodes were placed at 25% on the line between the head of the fibula to the lateral malleolus (Figure 3.1). To improve electrical contact the area was prepared for electrode placement by shaving and cleansing with an alcohol wipe. Root mean square was used for processing the EMG data using a moving window of 100 ms, each value in the signal is squared, averaged over the time interval, and then square rooted (Soderberg & Knutson, 2000).

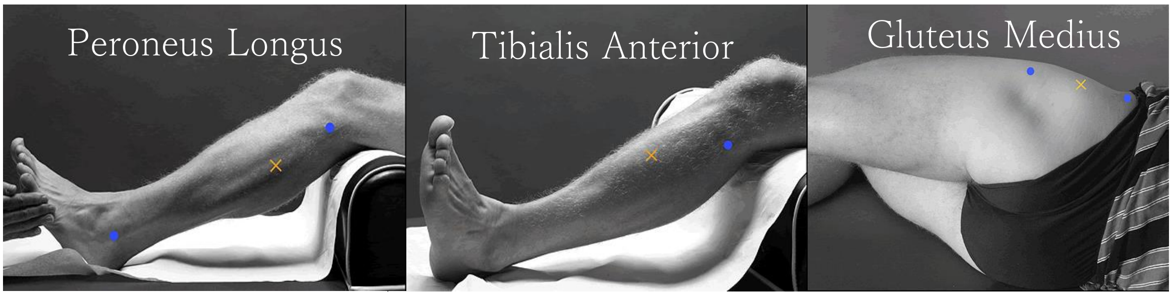


Figure 3.1 SENIAM guidelines for peroneus longus, tibialis anterior and gluteus medius electrode placements

3.5 Motion Analysis

Motion analysis data were recorded using an Owl Digital Real Time 10 camera system (Motion Analysis, Santa Rosa, California) sampling at 200 Hz. A rectangular capture volume measuring 1.5 m (width) by 2.1 m (height) by 3.5 m (length) was calibrated as per the manufacturer's instructions using an L-frame and T-bar wand. Passive reflective markers were attached to the participant using double-sided tape, in accordance with the Helen Hayes marker set (Davis et al., 1991) combined with the Oxford Foot Model (Stebbins et al., 2006; Wright et al., 2011) (Figure 3.2). Marker and electrode placement were performed by the same person for all participants. All data were inspected using

Cortex software (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) before importing into Visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland). Data were smoothed using a 6 Hz Butterworth filter. Relevant data as outlined in each individual chapter were imported into MATLAB R2015a (The Math Works, Natick, Massachusetts) for SPM analysis and SPSS (IBM SPSS Statistics for Windows, Version 24.0. Armonk, NY: IBM Corp) for discrete variable and regression analysis.

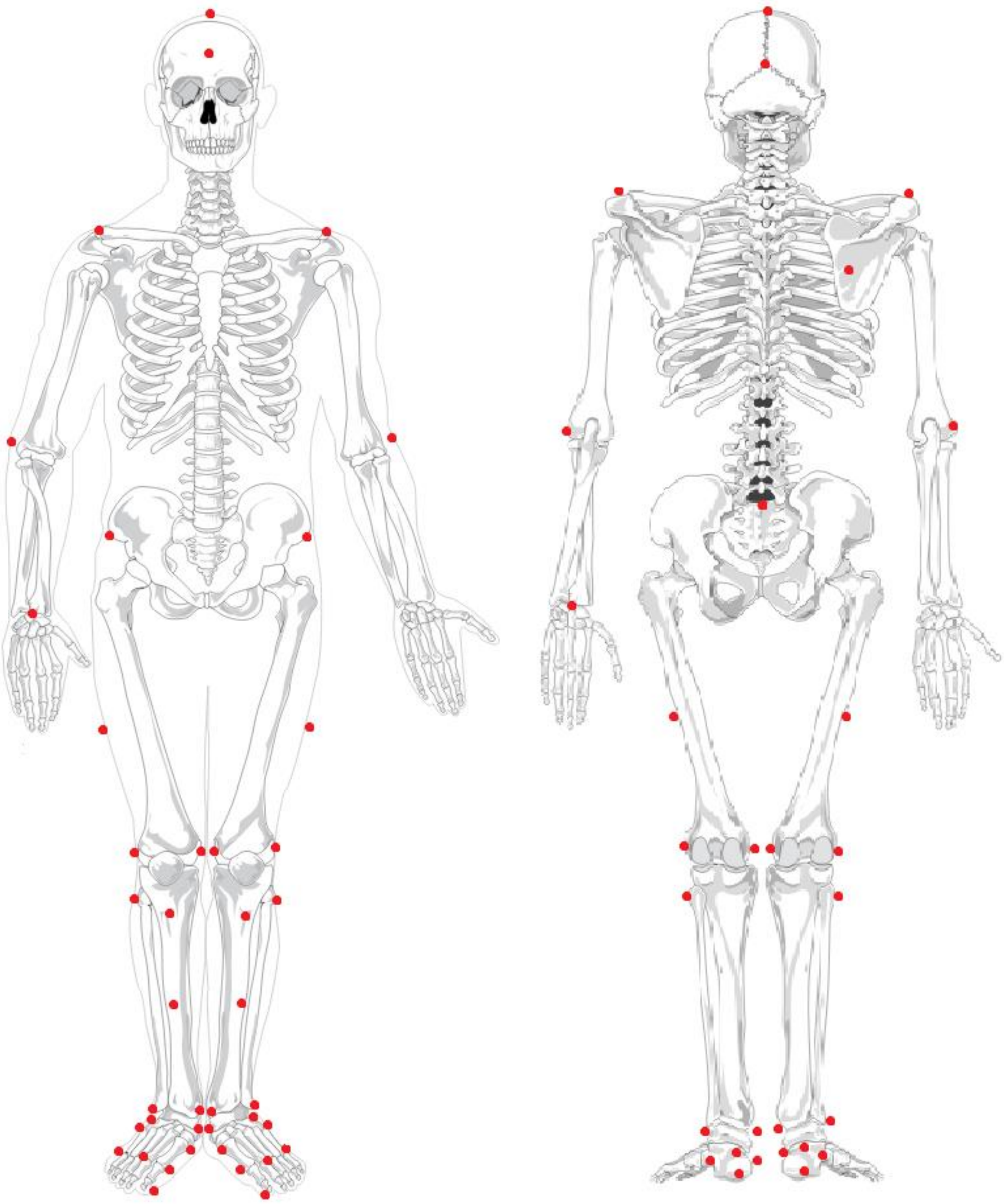


Figure 3.2 Combined Helen Hayes and Oxford Foot Model

3.6 Statistical Parametric Mapping

Due to the one dimensional nature of biomechanical data including time and kinematic variable or electromyographic data, previous research has suggested that reducing to zero-dimensional data (removing the time component) may result in focus bias or missing potential trends or areas of significance (Pataky et al., 2013). In Chapters 6.2, 7.2 and 8.2 statistical parametric mapping will be implemented to enable analysis of movement and muscle activation patterns. Raw data were first saved into text files for 101 normalised time nodes before importing into MATLAB R2015a (The Math Works, Natick, Massachusetts). On importing to MATLAB data were assessed for normality using a D'Agostino-Pearson's test. Coding was then written for each statistical test as outlined in each individual chapter.

The two samples *t*-test follows a 5-step process:

1) Mean fields were computed for each group across the selected time period under analysis
2) Standard deviations were then computed again for each group across the selected time period
3) *t*-test statistics were computed using the equation below:

$$t = \frac{Sample\ mean_{\ B}\ -\ Sample\ mean_{\ A}}{\sqrt{\frac{1}{Sample\ size}(St\ dev_{\ A}{}^{2} +\ St\ dev_{\ B}{}^{2})}}$$

4) Statistical inference was conducted using alpha (α) and random field theory t-distribution to compute critical threshold. Where SPM{t} > t_{critical} the null hypothesis can be rejected for the suprathreshold clusters
5) *p* values (probability that a completely random *n*D process will yield a particular result) were then computed for each cluster using cluster size and random field theory distributions for SPM{t} topology

Where data exceeds the threshold for normality statistical non-parametric mapping (SnPM) were conducted in addition to the normal SPM protocol. This was based on the recommendations of Todd Pataky (2015). It is stated that if the results of both tests do not differ qualitatively then the parametric approach's assumption of normality is a reasonable one.

3.7 Statistical Tests

Multiple *t*-tests were employed throughout this thesis as there were three key research questions under investigation:

1) Are there differences between a healthy control and the affected limb of individuals with ankle instability?
2) Is there compensation or differences between the affected and unaffected limb of those with ankle instability?
3) Is the "unaffected" limb of those with ankle instability comparable to an uninjured control?

Individual *t*-tests answer these questions. An ANOVA works on the following assumptions: Population are normally distributed, homogeneity of variance, data on parametric scale, scores in all groups are independent, scores in each group are not dependent on, not correlated with, or not taken from the same subjects as scores in any other group. Bonferroni corrections were not implemented within the analysis as these are thought to be over conservative and thus lead to an increase in type II error whereby hypotheses are incorrectly rejected.

Chapter 4.0 Reliability Testing

4.1 Camera Reliability

4.1.1 Introduction

The gold standard for motion analysis is three-dimensional motion analysis (Munro et al., 2012). Positional data is used to obtain angular displacements, angular velocities and angular accelerations (Melton et al., 2011). To highlight potential areas to target with rehabilitation and preventative strategies the information obtained from the motion analysis system must be accurate, reliable and valid. When measuring the reliability of camera systems most authors use human participants (Ferber, McClay Davis, Williams, & Laughton, 2002; Noonan et al., 2003; Tsushima, Morris, & McGinley, 2003). However, this creates an additional degree of variability within the data due to inconsistent movement strategies, movement speeds, marker placement and fatigue of the participant with multiple trials (McGinley, Baker, Wolfe, & Morris, 2009). To assess the accuracy and reliability of the camera system set up, a known marker distance and known marker velocity may provide more valid results. The aim of this study was to assess the coefficient of variation in multiple positions of the setup capture volume and across days.

4.1.2 Method

Testing was conducted across two consecutive days. Motion analysis data were recorded using an Owl Digital Real Time 10 camera system (Motion Analysis, Santa Rosa, California) sampling at 200 Hz. The motion analysis system was calibrated as per the manufacturer's instructions. Three passive reflective markers were attached to a known distance 30 cm linear actuator (Gimson Robotics, Bristol). Two markers were attached to the static components and one on the moving component of the actuator, as shown in

Figure 4.1. Markers remained fixed in place throughout the two days testing to eliminate marker placement errors.

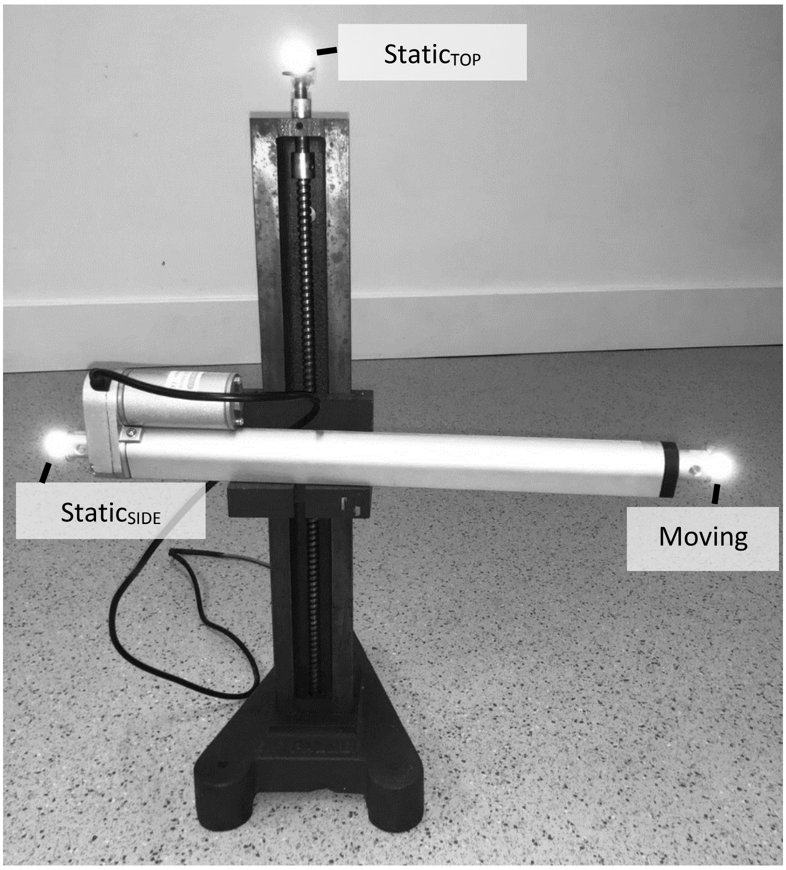


Figure 4.1 Linear actuator and marker placements

The recording volume was marked out onto the floor in a 3 x 6 square grid (50 cm x 50 cm). The actuator was positioned at 12 intersects of the grid as shown in Figure 4.2. The orientation was then rotated to the left (YL), right (Y) and front (X) creating a total of 36 locations for data collection. A 25-second duration capture of the linear actuator was recorded 3 times in each location and this was repeated for the subsequent days testing.

Data were inspected using Cortex software (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) before importing into Visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland). Data were smoothed using a 6 Hz Butterworth filter. Firstly a same day/between trial/ position comparison was performed. The peak distance from the moving marker and $static_{SIDE}$, the peak distance from the moving marker and $static_{TOP}$ and the peak velocity of the moving marker were exported and the mean and standard deviation (SD) calculated for each position across the three trials for day one. Coefficient of Variation (CV) was calculated using the formula Coefficient of Variation = (Standard Deviation / Mean) * 100.

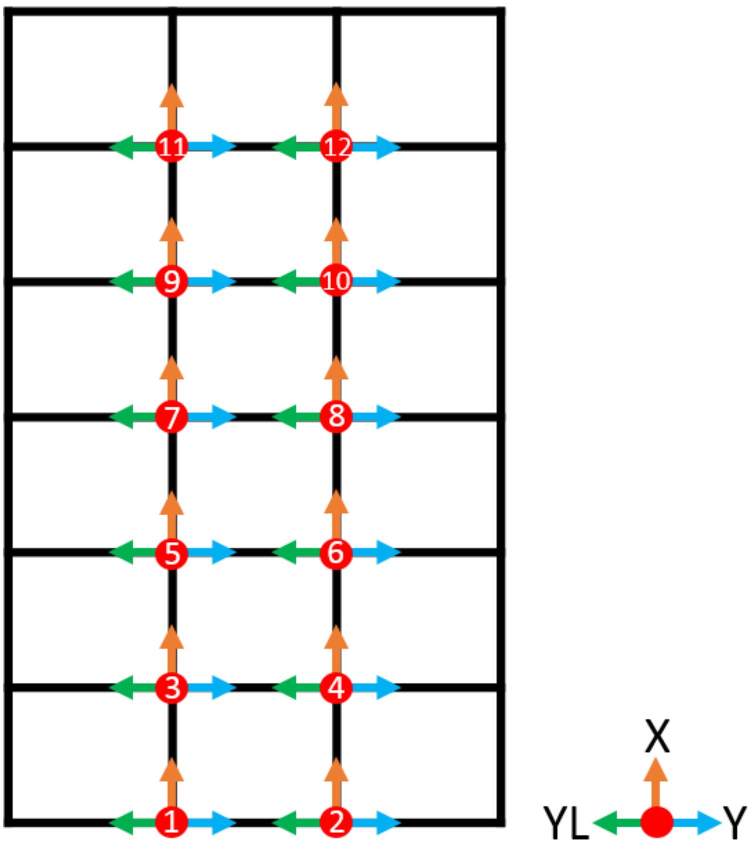


Figure 4.2 Showing linear actuator locations and orientations for data collection.

4.1.3 Results

Day 1 – Comparison of Position - Coefficient of variation were consistently low with maximum values being 0.16% (moving – $static_{SIDE}$), 0.11% (moving – $static_{TOP}$) and 1.86% (moving velocity) (Table 4.1).

Across Day Testing - Coefficient of variation was consistently low with maximum values being 0.32 % (moving – $static_{SIDE}$), 0.30 % (moving – $static_{TOP}$) and 9.58% (moving velocity) (Table 4.2).

Table 4.1 Means, standard deviations and coefficients of variation for moving-static$_{SIDE}$, moving-static$_{TOP}$ and velocity for each location within the capture volume on day 1

	Moving – Static$_{SIDE}$ (meters)			Moving – Static$_{TOP}$ (meters)			Velocity (m/s)		
	Mean	SD	CV (%)	Mean	SD	CV (%)	Mean	SD	CV (%)
X1	0.30	0.00	0.01	0.30	0.00	0.03	0.03	0.00	0.14
X2	0.30	0.00	0.05	0.30	0.00	0.01	0.03	0.00	0.25
X3	0.30	0.00	0.00	0.30	0.00	0.02	0.03	0.00	0.19
X4	0.31	0.00	0.05	0.31	0.00	0.00	0.03	0.00	0.15
X5	0.31	0.00	0.06	0.31	0.00	0.06	0.03	0.00	0.46
X6	0.31	0.00	0.02	0.31	0.00	0.04	0.03	0.00	0.58
X7	0.31	0.00	0.02	0.31	0.00	0.00	0.05	0.00	1.41
X8	0.31	0.00	0.02	0.30	0.00	0.07	0.03	0.00	0.50
X9	0.31	0.00	0.00	0.31	0.00	0.05	0.04	0.00	0.44
X10	0.30	0.00	0.04	0.30	0.00	0.00	0.04	0.00	1.68
X11	0.31	0.00	0.02	0.31	0.00	0.00	0.04	0.00	1.29
X12	0.31	0.00	0.04	0.31	0.00	0.00	0.04	0.00	1.86
Y1	0.30	0.00	0.05	0.30	0.00	0.05	0.03	0.00	0.44
Y2	0.31	0.00	0.00	0.31	0.00	0.00	0.03	0.00	0.32
Y3	0.30	0.00	0.16	0.31	0.00	0.00	0.03	0.00	0.21
Y4	0.30	0.00	0.05	0.31	0.00	0.02	0.03	0.00	0.20
Y5	0.31	0.00	0.04	0.31	0.00	0.01	0.03	0.00	0.09
Y6	0.30	0.00	0.07	0.30	0.00	0.06	0.03	0.00	0.12
Y7	0.31	0.00	0.01	0.31	0.00	0.01	0.03	0.00	0.30
Y8	0.31	0.00	0.05	0.31	0.00	0.06	0.03	0.00	0.45
Y9	0.30	0.00	0.07	0.30	0.00	0.06	0.03	0.00	0.15
Y10	0.31	0.00	0.04	0.31	0.00	0.04	0.03	0.00	0.48
Y11	0.31	0.00	0.07	0.31	0.00	0.07	0.03	0.00	0.28
Y12	0.31	0.00	0.10	0.31	0.00	0.11	0.04	0.00	0.62
YL1	0.30	0.00	0.00	0.30	0.00	0.01	0.03	0.00	0.23
YL2	0.31	0.00	0.01	0.31	0.00	0.01	0.03	0.00	0.27
YL3	0.31	0.00	0.02	0.31	0.00	0.01	0.03	0.00	0.20
YL4	0.30	0.00	0.02	0.30	0.00	0.02	0.03	0.00	0.24
YL5	0.31	0.00	0.03	0.31	0.00	0.01	0.03	0.00	0.19
YL6	0.31	0.00	0.02	0.31	0.00	0.02	0.03	0.00	0.19
YL7	0.31	0.00	0.02	0.31	0.00	0.01	0.03	0.00	0.53
YL8	0.31	0.00	0.02	0.31	0.00	0.00	0.03	0.00	0.56
YL9	0.31	0.00	0.08	0.31	0.00	0.08	0.03	0.00	0.31
YL10	0.31	0.00	0.04	0.31	0.00	0.05	0.03	0.00	0.24
YL11	0.30	0.00	0.01	0.30	0.00	0.01	0.03	0.00	0.14
YL12	0.31	0.00	0.05	0.31	0.00	0.04	0.03	0.00	0.38

Table 4.2 Means, standard deviations and coefficients of variation for moving-static$_{SIDE}$, moving-static$_{TOP}$ and velocity across days

	Moving – Static (meters)			Moving – Top (meters)			Velocity (m/s)		
	Day 1	Day 2	Day 3	Day 1	Day 2	Day 3	Day 1	Day 2	Day 3
Mean	0.31	0.31	0.31	0.31	0.31	0.31	0.03	0.03	0.03
SD	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00	0.00
CV	0.26	0.23	0.32	0.26	0.23	0.30	9.58	6.55	7.65

4.1.4 Conclusion

To improve reliability, each movement under investigation throughout this thesis was performed at the same point within the capture volume to optimise the location and limit variability in results. Each participant completed testing in one day to limit variation in results due to differences in camera setup and marker placement. Greater deviation is observed in velocity so care should be taken with this.

4.2 Marker Set Reliability

4.2.1 Introduction

Previous research in the University of Hertfordshire laboratory has observed very good to excellent reliability for the use of the Helen Hayes marker set during jumping and landing – sagittal plane motion ICC's hip = 0.92, knee = 0.95 and frontal plane hip = 0.67 and knee = 0.67 (Hunter, 2017). Similarly, the Oxford Foot Model has also been shown to be repeatable for inter and intratester repeatability (Carson et al., 2001; McCahill, Stebbins, Koning, Harlaar, & Theologis, 2018; Stebbins et al., 2006; van Hoeve et al., 2015). Variation in results could be due to variations in individual movement mechanics, marker placement error or camera variability. This study aims to remove two of these sources of variability by reporting inter-marker and angular displacements known values across a fixed model across multiple days and recording positions.

4.2.2 Method

Prior to testing with the combined Oxford Foot Model and Helen Hayes marker set authors tested the reliability of the camera set up. Testing was conducted across two consecutive days. Motion analysis data were recorded using an Owl Digital Real Time 10 camera system (Motion Analysis, Santa Rosa, California) sampling at 200 Hz. A combined Oxford Foot Model and Helen Hayes marker set were used in this study as outlined in general methods (3.5) for the right leg. This marker set screwed into position on a fixed joint skeleton to ensure the spheroid distance maintained the same between each marker. Markers remained fixed in place throughout the two days testing to eliminate marker placement errors.

The recording volume was marked out onto the floor in a 3 x 6 square grid (50 cm x 50 cm). The skeleton was placed in the centre of each point of the grid (Figure 4.3). Three 10 second recordings were taken in each position. This protocol was repeated the following day.

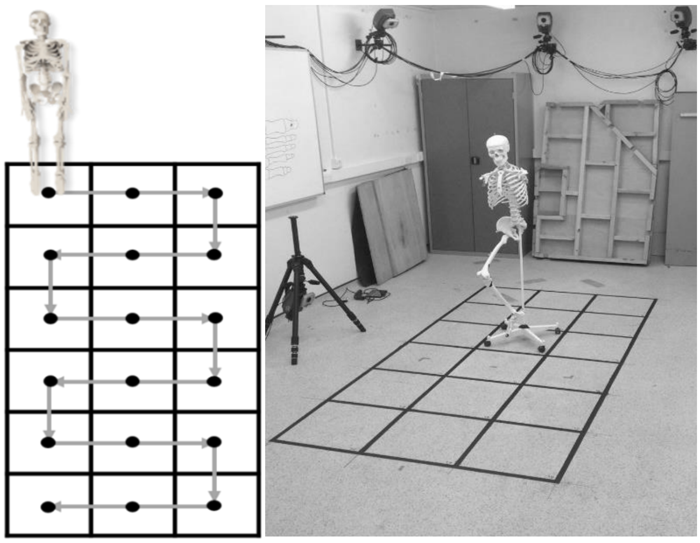


Figure 4.3 Skeleton placement and locations for data

Six inter-marker distances (Figure 4.4) and an angular position were selected at random for analysis. Means and standard deviations were reported for inter-marker distances and hip angle in the sagittal, frontal and transverse planes. Coefficient of variations were also calculated using the formula Coefficient of Variation = (Standard Deviation / Mean) * 100.

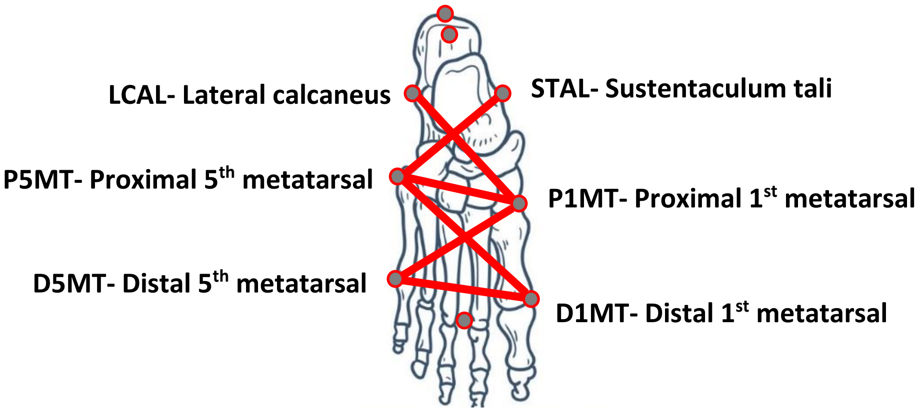


Figure 4.4 Inter-marker distances

4.2.3 Results

Coefficients of variation calculated for inter-marker distances for the three trials in each position across the capture volume showed low CV between positions (Table 4.3).

Table 4.3 Means, standard deviations (cm) and coefficients of variation for inter-marker distances for the three trials in each position on day 1

	D1MT_D5MT			P1MT_P5MT			D1MT_P5MT			D5MT_P1MT			P1MT_LCAL			P5MT_STAL		
	Mean	SD	CV (%)	Mean	SD	CV (%)	Mean	SD	CV (%)	Mean	SD	CV (%)	Mean	SD	CV (%)	Mean	SD	CV (%)
Position 1	7.91	0.00	0.01	7.79	0.00	0.00	9.57	0.00	0.02	8.16	0.00	0.00	10.89	0.00	0.02	8.06	0.00	0.00
Position 2	7.83	0.00	0.00	8.04	0.01	0.08	9.48	0.00	0.00	8.42	0.00	0.01	11.11	0.01	0.09	7.93	0.00	0.00
Position 3	7.79	0.01	0.14	7.90	0.00	0.01	9.33	0.00	0.01	8.41	0.00	0.04	11.03	0.00	0.01	8.02	0.00	0.00
Position 4	8.00	0.00	0.02	7.59	0.00	0.00	9.44	0.00	0.01	8.24	0.00	0.02	10.86	0.00	0.00	7.77	0.00	0.00
Position 5	7.95	0.00	0.00	8.03	0.00	0.01	9.62	0.00	0.00	8.41	0.00	0.00	10.90	0.00	0.01	8.16	0.00	0.00
Position 6	7.94	0.00	0.03	8.02	0.01	0.18	9.56	0.01	0.12	8.41	0.00	0.02	11.12	0.00	0.01	7.92	0.01	0.10
Position 7	7.89	0.00	0.00	8.10	0.00	0.04	9.75	0.01	0.05	8.34	0.00	0.00	10.59	0.00	0.00	8.12	0.00	0.01
Position 8	7.98	0.00	0.00	7.78	0.00	0.04	9.32	0.00	0.01	8.37	0.00	0.04	11.02	0.00	0.01	7.87	0.00	0.00
Position 9	7.92	0.00	0.04	7.97	0.00	0.01	9.59	0.00	0.00	8.43	0.00	0.03	10.76	0.00	0.00	7.91	0.00	0.01
Position 10	7.92	0.02	0.19	7.99	0.00	0.01	9.69	0.02	0.16	8.34	0.00	0.02	10.70	0.00	0.04	8.13	0.00	0.01
Position 11	7.95	0.01	0.11	7.88	0.00	0.00	9.52	0.01	0.07	8.37	0.00	0.00	10.92	0.00	0.00	7.87	0.00	0.00
Position 12	7.76	0.00	0.01	7.85	0.00	0.03	9.40	0.00	0.01	8.16	0.00	0.03	11.02	0.00	0.01	8.05	0.00	0.03
Position 13	7.92	0.00	0.04	7.90	0.02	0.22	9.52	0.01	0.11	8.34	0.01	0.13	10.81	0.01	0.13	8.02	0.00	0.00
Position 14	7.88	0.00	0.00	7.96	0.01	0.14	9.47	0.00	0.00	8.30	0.01	0.14	10.87	0.03	0.25	8.23	0.00	0.00
Position 15	7.98	0.01	0.07	7.84	0.01	0.07	9.55	0.01	0.09	8.20	0.01	0.07	10.86	0.02	0.16	8.05	0.00	0.03
Position 16	7.88	0.02	0.23	7.87	0.02	0.25	9.33	0.01	0.07	8.32	0.01	0.09	11.04	0.00	0.02	8.14	0.00	0.05
Position 17	7.73	0.00	0.03	7.89	0.00	0.02	9.50	0.01	0.07	8.27	0.01	0.06	11.09	0.00	0.01	7.90	0.00	0.01
Position 18	7.96	0.00	0.05	7.77	0.00	0.00	9.36	0.01	0.11	8.40	0.00	0.00	10.95	0.00	0.00	8.21	0.00	0.05

Coefficients of variation were calculated for inter-marker distances for day 1 and day 2 (Table 4.4) these show low dispersion of data about the mean with a peak value of 2.64 % and a lowest value of 0.78 %.

Table 4.4 Means, standard deviations (cm) and coefficients of variation for inter-marker distances for day 1 and day 2

	DAY 1			DAY 2		
	Mean	SD	CV (%)	Mean	SD	CV (%)
D1MT-D5MT	7.90	0.08	0.96	7.92	0.09	1.10
P1MT-P5MT	7.90	0.12	1.52	7.82	0.08	1.07
D1MT_P5MT	9.50	0.12	1.28	9.46	0.11	1.19
D5MT_P1MT	8.33	0.09	1.04	8.26	0.06	0.78
P1MT_LCAL	10.92	0.14	1.30	10.73	0.28	2.64
P5MT_STAL	8.02	0.13	1.61	8.09	0.08	1.03

Coefficients of variation calculated for the hip sagittal, frontal and transverse plane on day 1 and day 2 showed a peak CV in the transverse plane for day 2 of 3.12 % and a minimum value of 0.33 % on day 1 in the sagittal plane (Table 4.5).

Table 4.5 Means, standard deviations (degrees) and coefficients of variation for sagittal, frontal and transverse plane hip angles between days

	Sagittal plane			Frontal plane			Transverse plane		
	Mean	SD	CV (%)	Mean	SD	CV (%)	Mean	SD	CV (%)
Day 1	66.68	0.22	0.33	-12.77	0.16	1.22	-13.95	0.29	2.11
Day 2	66.60	0.26	0.39	-12.85	0.18	1.41	-13.52	0.42	3.12

4.2.4 Conclusion

Low CV was observed between positions in the calibrated capture volume. To improve reliability each movement under investigation will be performed at the same point within the capture volume. Each participant will also complete all testing in one testing day to limit variation due to repeat marker placement and differences in camera set up.

Chapter 5.0 Pilot Testing

5.1 Dominant/Non-Dominant Comparison During Walking, Landing and Cutting Manoeuvres

5.1.1 Introduction

Lateral ankle sprains are not exclusive to either the dominant or non-dominant limb (Woods et al., 2003; Yeung et al., 1994). When recruiting participants it is important to be aware of the possible differences that may occur between the dominant and the non-dominant limb to guide methods for future studies. The aim of this study was to compare full time series movement analysis between the dominant and the non-dominant limb during walking, landing and cutting movements.

5.1.2 Method

Participants

Eighteen (14 male, 4 female) healthy controls (age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg) participated in this study. Ethical approval was granted by the university's ethics committee prior to testing. Written informed consent was obtained and a health screen questionnaire was completed prior to participation.

Participants were included in the study if they were aged 18-35 and participated in sport a minimum of twice a week (minimum of 30 minutes per session). Participants were excluded if any of the following applied; existing acute lower limb injury in past 3 months, use of prescribed orthotics, lower extremity biomechanical abnormality, balance or motion disorders, history of fracture requiring realignment or history of lower extremity surgery. The dominant limb was determined by asking participants which limb they would use to kick a ball.

Walking trials and analysis were conducted as per Chapter 6.2.2 where participants walked barefoot 3.5 m before data were collected (Najafi, Miller, Jarrett, & Wrobel, 2010) and proceeded for 7 m across the walkway at their normal walking speed through the calibrated capture volume. Pace was not controlled, as this was deemed to be unnatural and has been previously shown to impact on stride time variability due to increased central nervous system involvement (Springer & Gottlieb, 2017). Landing trials and analysis were conducted as per Chapter 7.2.2 - participants performed three barefoot single leg drop landings onto each limb from a 30 cm high box (Kunugi et al., 2017) onto a flat stable laboratory floor with 1-minute rest between trials. The order of trials was randomised to minimise the effect of fatigue. Individuals were asked to hop forward off the box onto the floor in front and maintain balance for 3 seconds whilst looking straight forward. No instruction was given to participants regarding arm position during the landing manoeuvre in order to observe an unmodified landing position. Cutting trials and analysis were conducted as per chapter 8.2.2. Participants were instructed to stand with feet shoulder width apart with weight equally distributed over both feet on a 30 cm box (Kunugi et al., 2017). They were instructed to jump two-footed forward off the box landing two-footed before performing a 90° cut.

Kinematic data were exported for forefoot-hindfoot angle (FFHFA), forefoot-tibia angle (FFTBA), hindfoot-tibia angle (HFTBA), hip, knee and trunk angles in the sagittal, frontal and transverse planes of motion. Data were analysed using SPM in MATLAB (SPM1D open-source package, spm1d.org). Normality was tested using the D'Agostino-Pearson's test. Dominant limb was compared to the non-dominant limb using a paired-samples *t*-test ($\alpha = 0.05$).

5.1.3 Results

5.1.3.1 Walking –Kinematics- Heel strike to toe-off

Paired sample *t*-tests were conducted to compare the dominant and non-dominant limb during the HS to toe off phase of walking. A significant difference was observed between HFTBA transverse plane motion at 83-88% of the HS to toe-off phase (p = 0.049). A significant difference was also observed between the dominant and non-dominant limb at 0-36% of the phase in hip transverse plane motion (p = 0.014). When observing knee transverse plane motion significant differences were observed at 0-14% (p = 0.042) and 20-99% (p = <0.001). Significant differences were observed in trunk transverse plane motion across the entire HS to toe-off phase of gait (0-100%) (p = 0.008). No other significant differences were observed between the dominant and non-dominant limbs of the healthy control group during the HS to toe-off phase of gait (Figure 5.1and Figure 5.2).

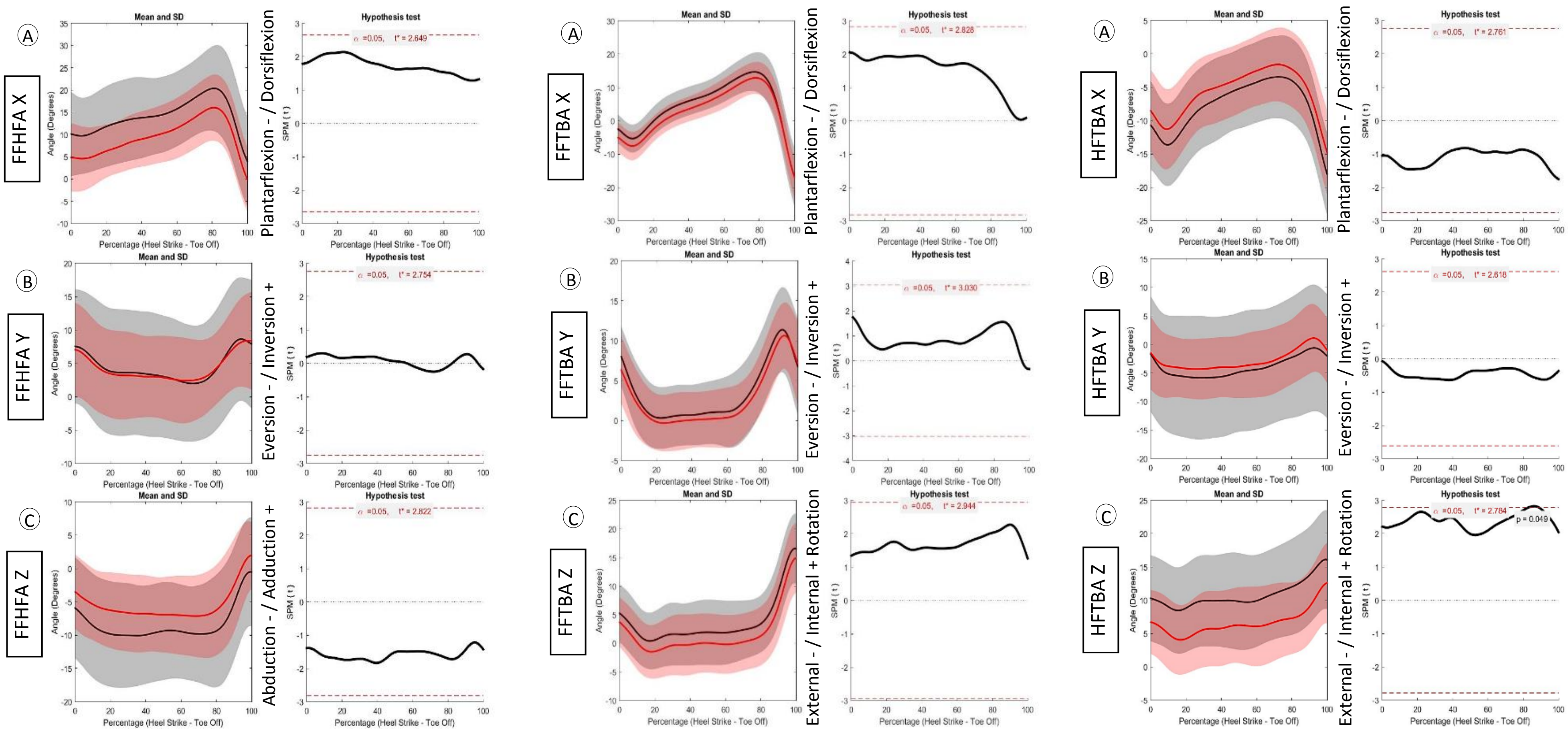


Figure 5.1 Walking - FFHFA, FFTBA and HFTBA displacement (x, y, z) Heel strike – toe off - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

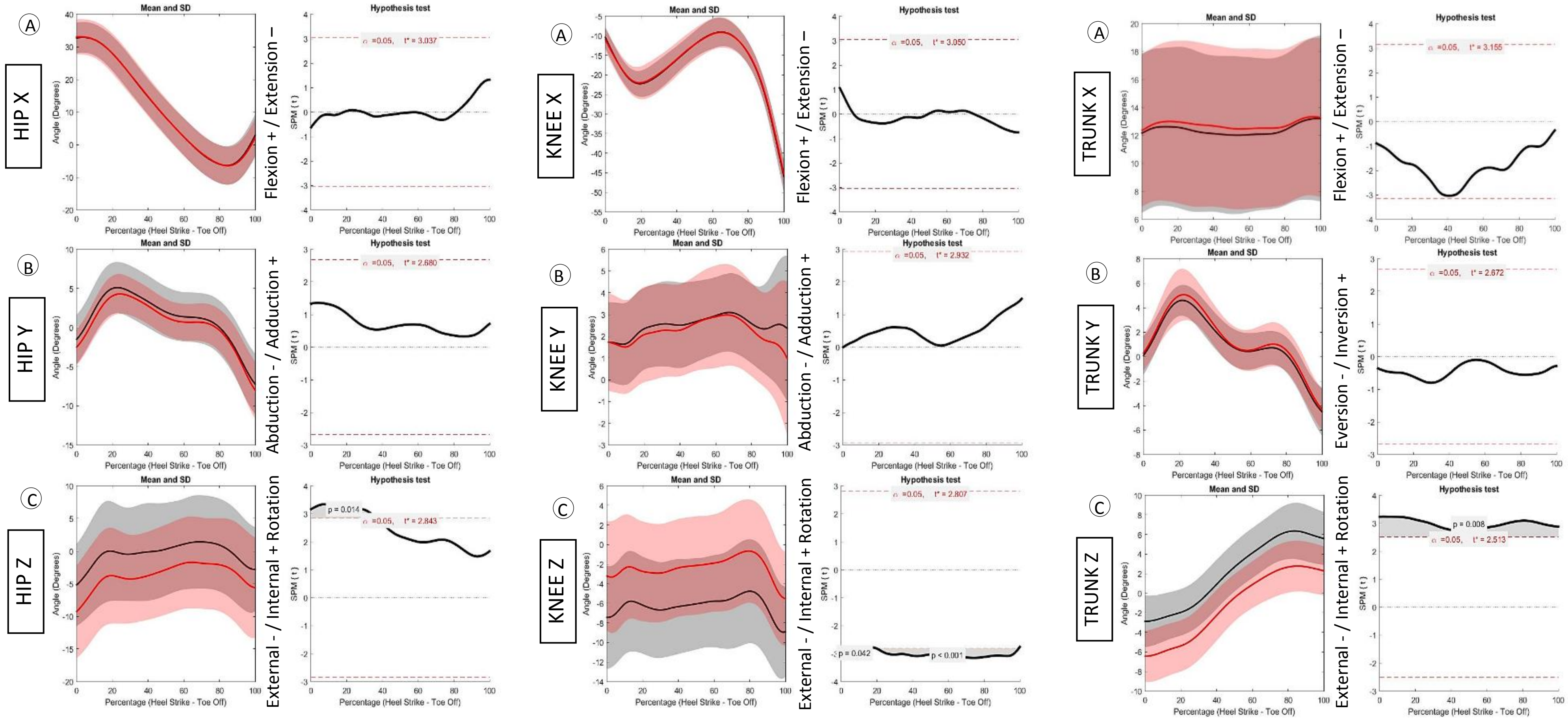


Figure 5.2 Walking - Hip, Knee and Trunk displacement (x, y, z) Heel strike – toe off - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

5.1.3.2 Walking –Kinematics - Toe off to Heel strike

Paired sample *t*-tests revealed a significant difference in HFTBA transverse plane motion at 89-96% of the toe off to HS phase (p = 0.048). A significant difference was also observed in knee transverse plane motion at 34-46% (p = 0.044) and 61-100% (p = 0.015) of the toe off to HS phase. When investigating trunk transverse plane motion significant differences were observed between the dominant and non-dominant limbs throughout the entire phase (0-100%) (p = 0.023). No other significant differences were observed in walking kinematics between the dominant and non-dominant limbs (Figure 5.3 and Figure 5.4Figure 5.4).

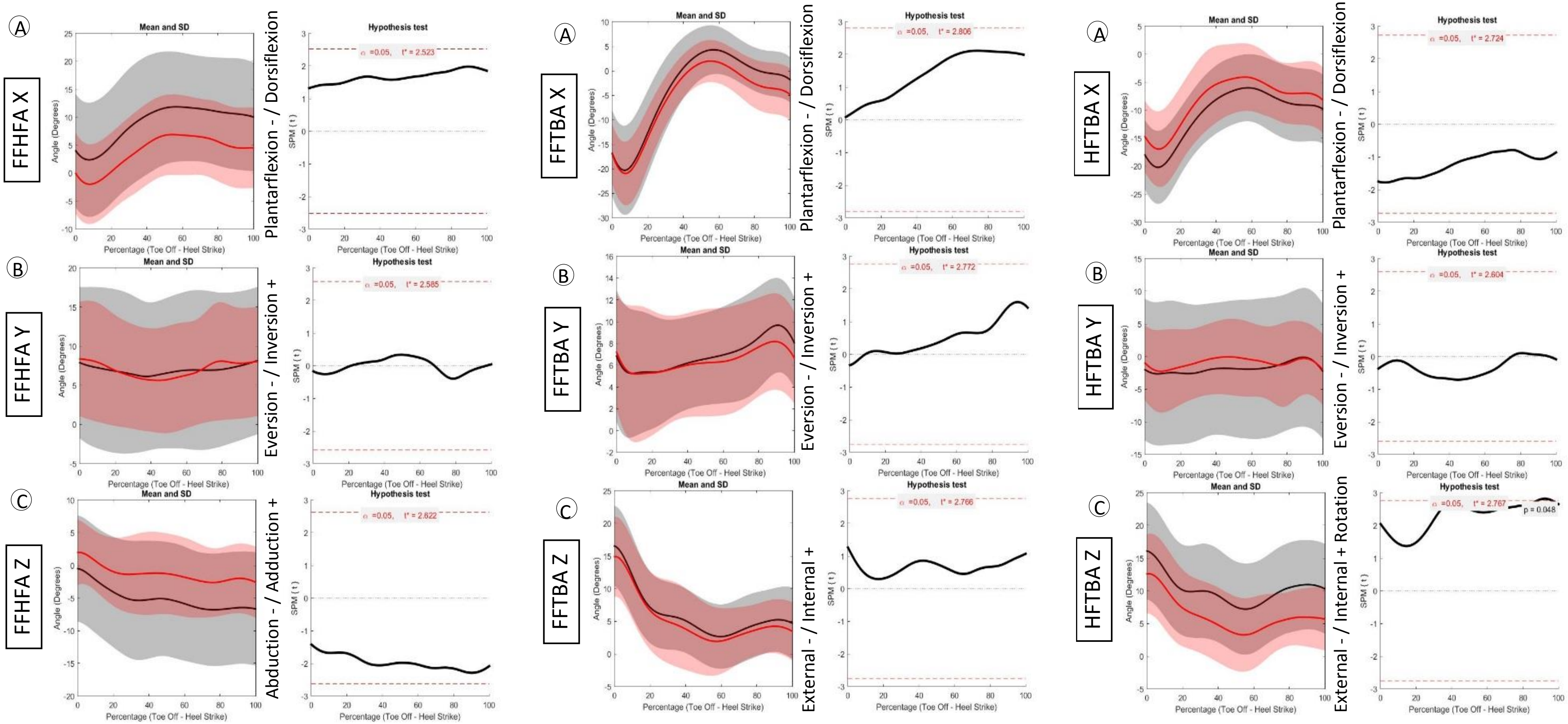


Figure 5.3 Walking - FFHFA, FFTBA and HFTBA displacement (x, y, z) Toe off - Heel strike- mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

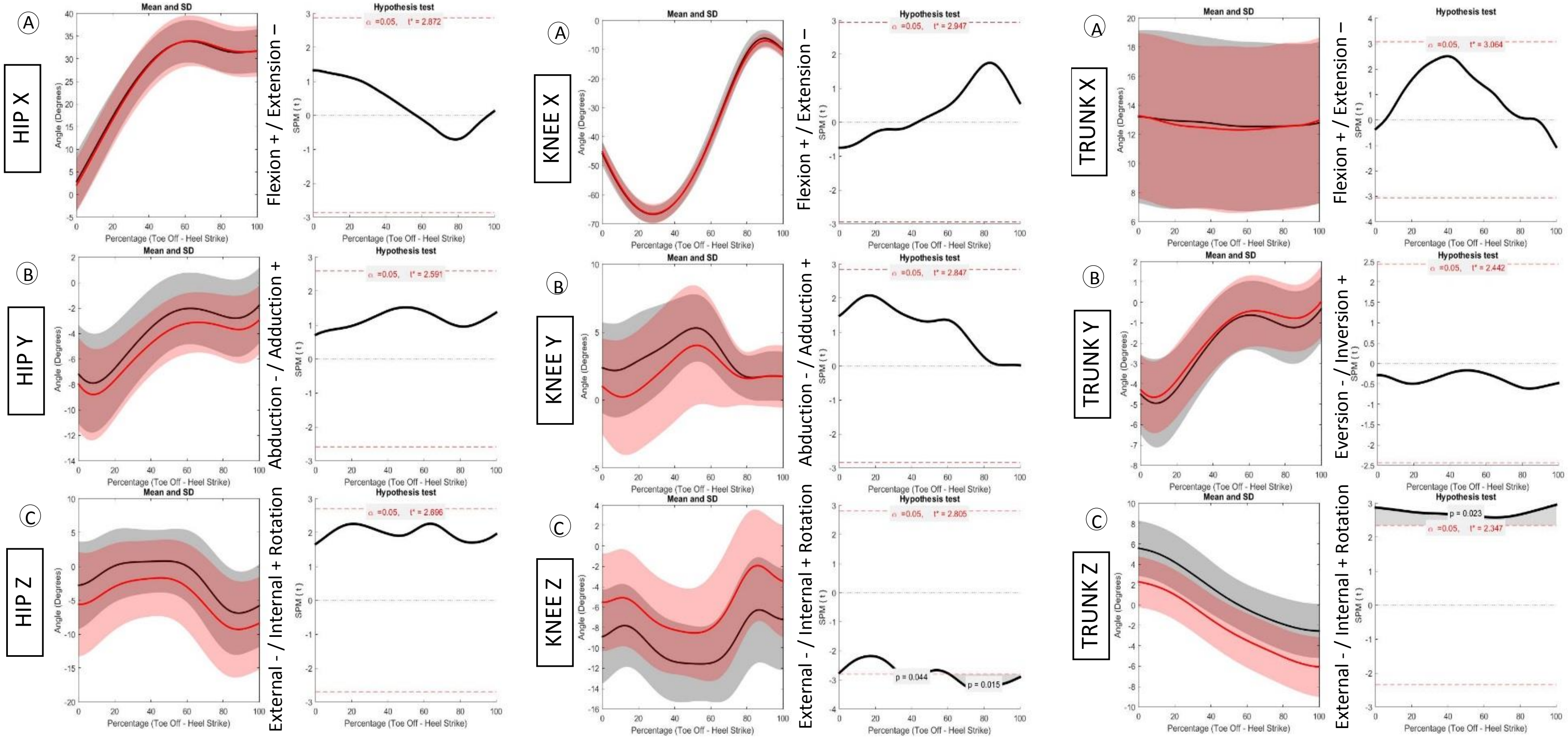


Figure 5.4 Walking - Hip, Knee and Trunk displacement (x, y, z) Toe off - Heel strike- mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

5.1.3.3 Landing –Kinematics- 200 ms prior to initial contact to initial contact

Paired samples *t*-tests conducted to compare landing kinematics during the 200 ms prior to IC to IC (pre-landing) phase showed a significant difference in hip frontal plane motion from 0-11% (p = 0.049). A significant difference was also observed in knee transverse plane motion from 0-58% (p = 0.011) of the pre-landing phase. No other significant differences were observed in kinematics between the dominant and non-dominant limb (Figure 5.5 and Figure 5.6).

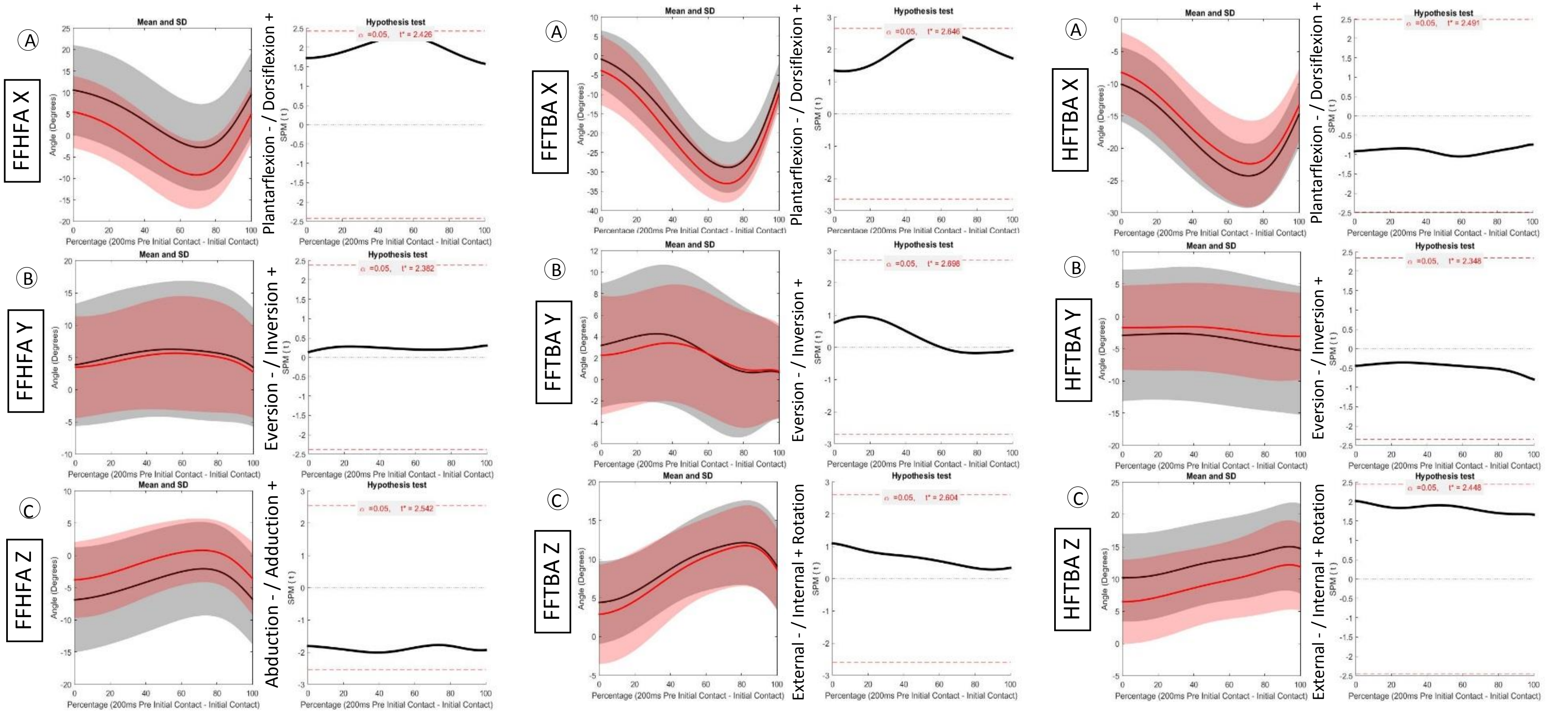


Figure 5.5 Landing - FFHFA, FFTBA and HFTBA displacement (x, y, z) 200 ms prior to initial contact to initial contact - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

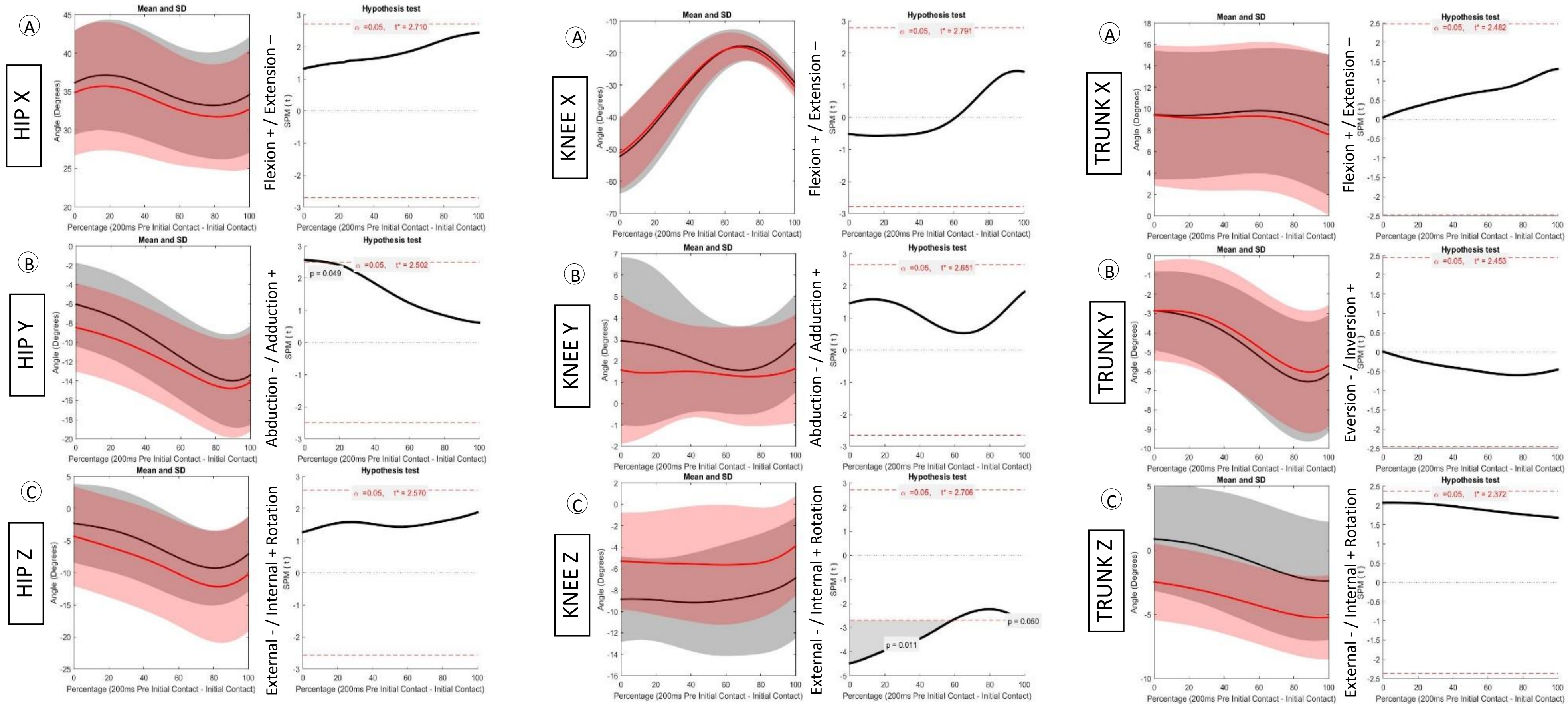


Figure 5.6 Landing - Hip, Knee and Trunk displacement (x, y, z) 200 ms prior to initial contact to initial contact - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

5.1.3.4 Landing –Kinematics- Initial contact to 200 ms post initial contact

During the impact phase of landing, paired samples *t*-tests showed significant differences between knee transverse plane kinematics at 0-20% (p = 0.046) and 76-100% (p = 0.045) between the dominant and non-dominant limb. Significant differences were also observed in trunk frontal plane motion at 82-100% of the impact phase of landing (p = 0.046). No other significant differences were observed between the dominant and non-dominant limb (Figure 5.7 and Figure 5.8).

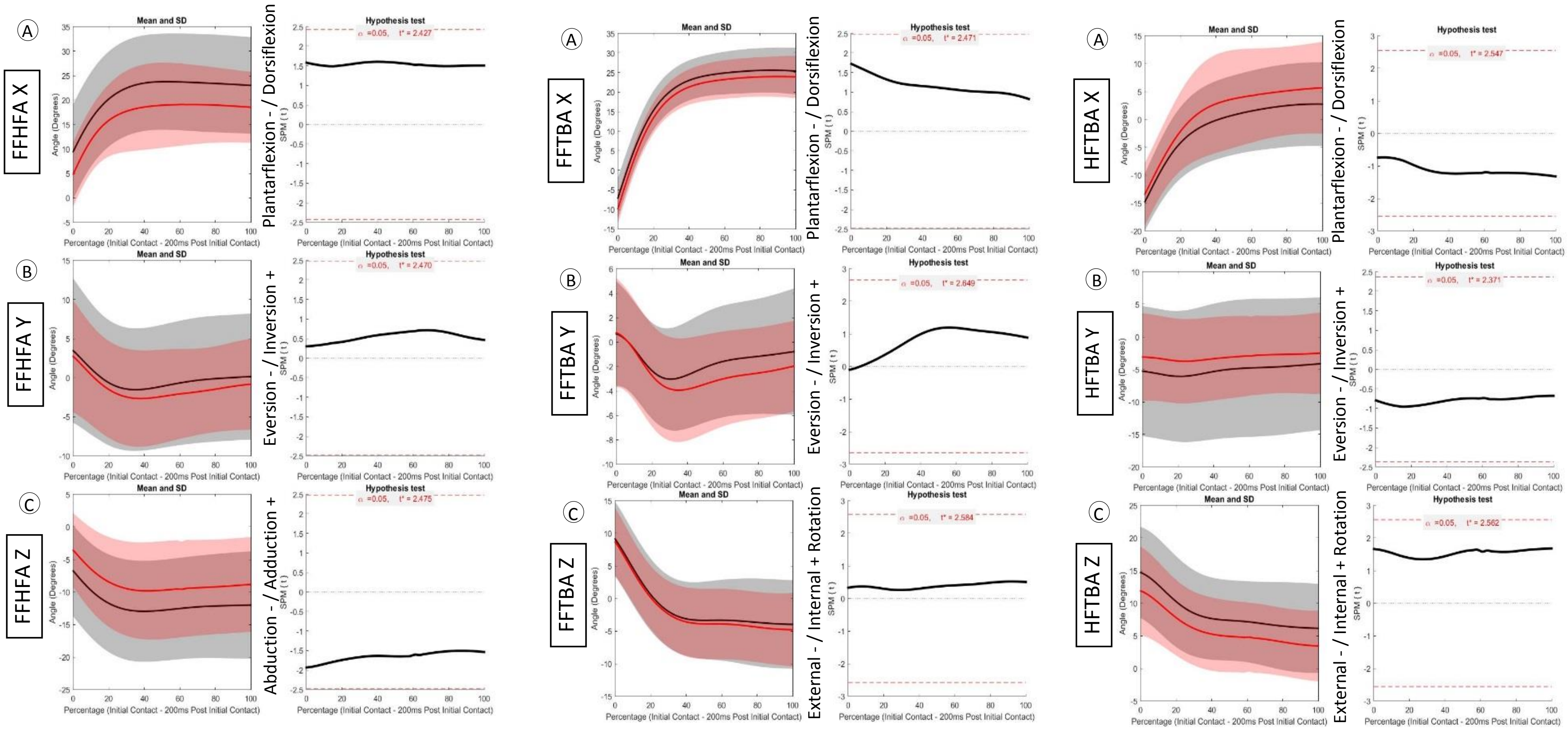


Figure 5.7 Landing - FFHFA, FFTBA and HFTBA displacement (x, y, z) Initial contact to 200 ms post initial contact - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

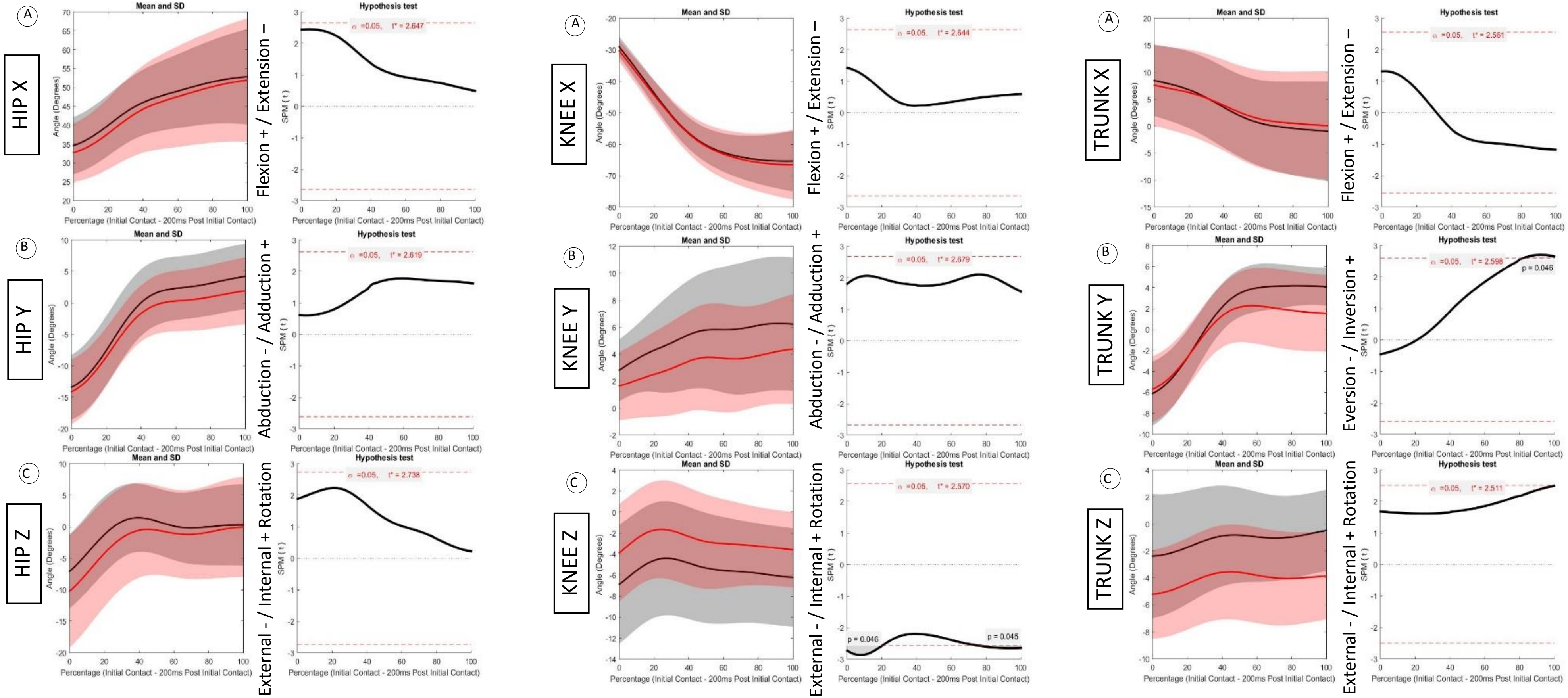


Figure 5.8 Landing - Hip, Knee and Trunk displacement (x, y, z) Initial contact to 200 ms post initial contact - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

5.1.3.5 Cutting –Kinematics- 200 ms prior to initial contact to foot strike

During cutting, paired sample *t*-tests revealed significant differences in transverse plane FFTBA, HFTBA and trunk kinematics between the dominant and non-dominant limb comparisons during the 200 ms prior to IC to IC. Differences in FFTBA transverse plane motion were observed at 46-82% (p = 0.014), differences in HFTBA at 22-77% (p = 0.032) and in the trunk across the entire phase (0-100%) (p = 0.014). No other significant differences were observed in any of the three planes of motion between limbs (Figure 5.9 and Figure 5.10).

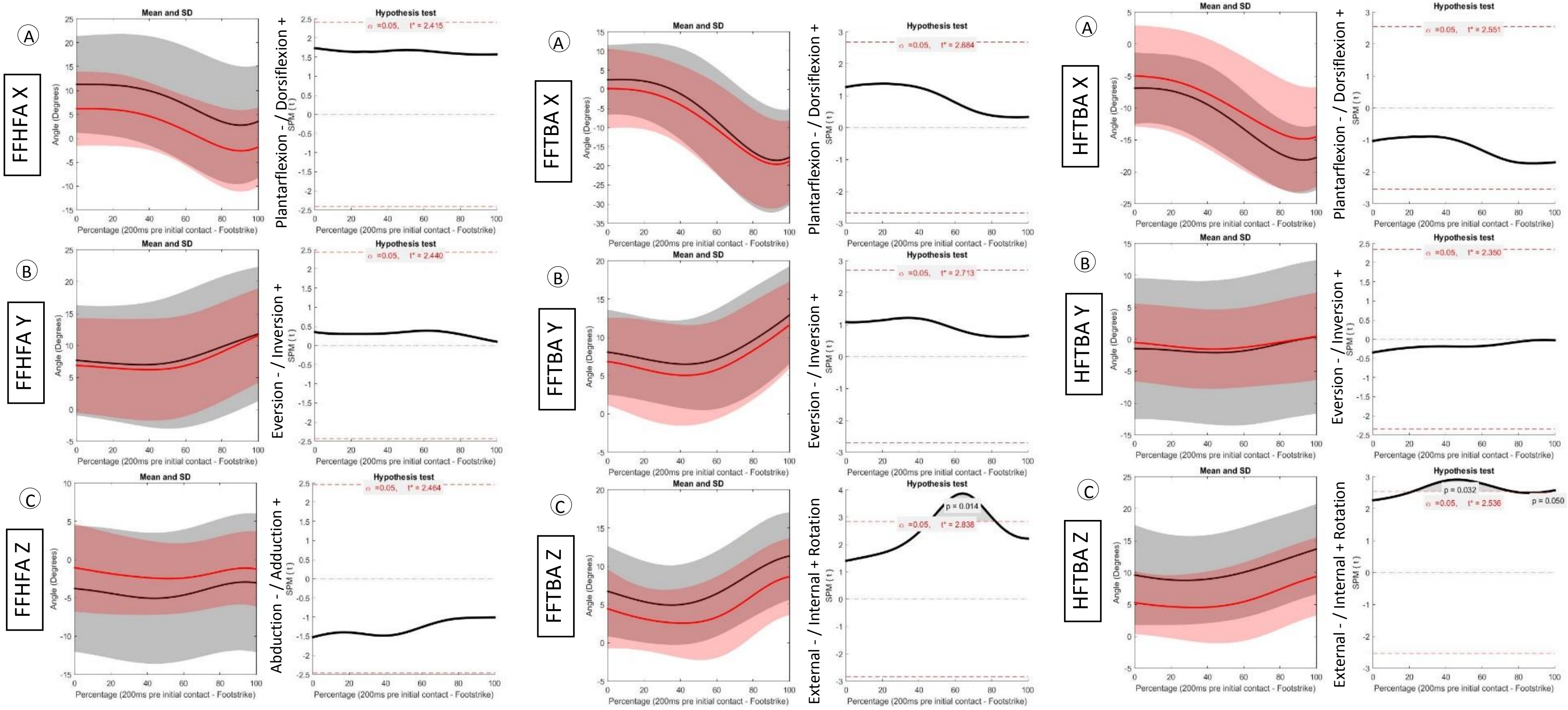


Figure 5.9 Cutting - FFHFA, FFTBA and HFTBA displacement (x, y, z) 200 ms prior to initial contact to foot strike- mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

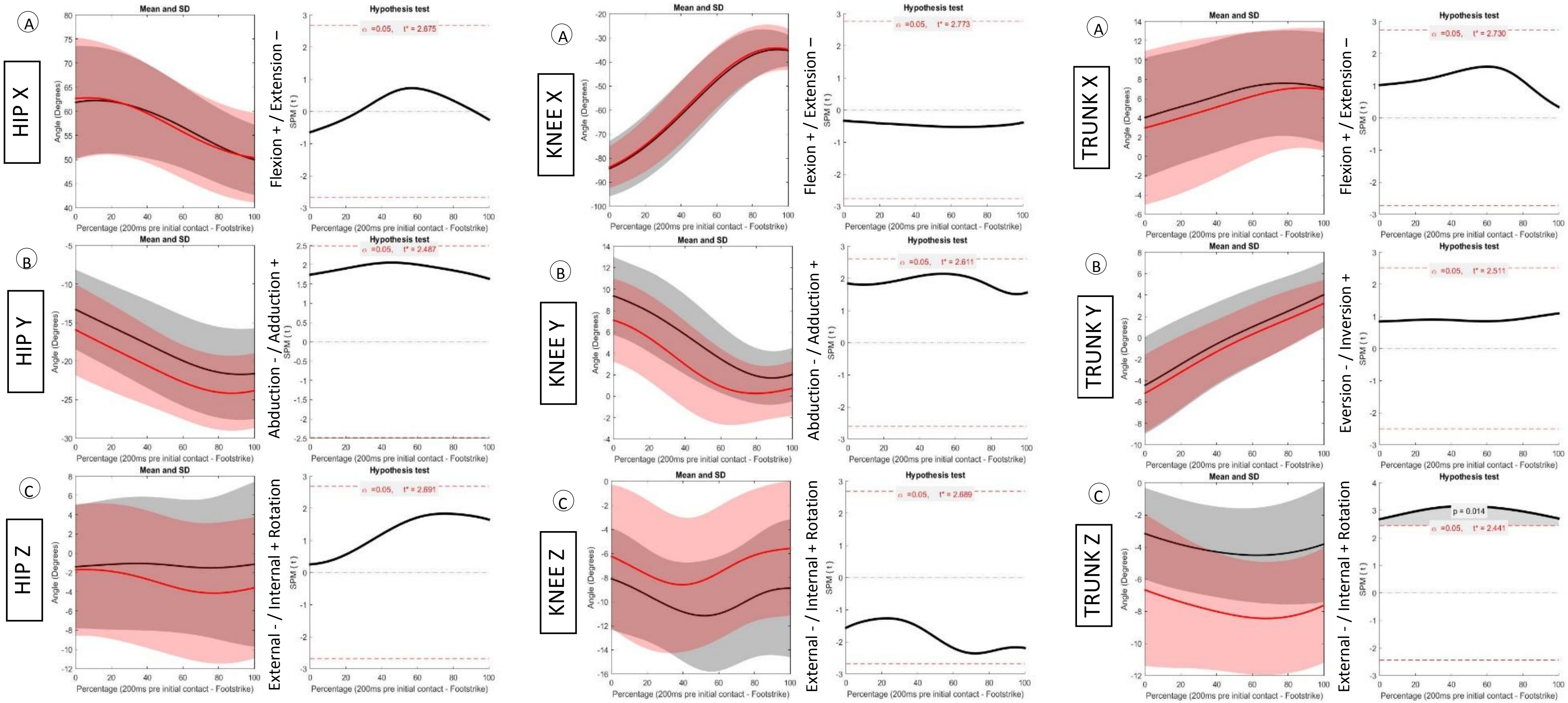


Figure 5.10 Cutting - Hip, Knee and Trunk displacement (x, y, z) 200 ms prior to initial contact to foot strike- mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

5.1.3.6 Cutting –Kinematics- Foot strike to toe-off

Paired sample *t*-tests revealed a significant difference in FFTBA frontal and transverse plane motion at 84-89% ($p = 0.047$) and 98-100% ($p = 0.049$) respectively. On analysis of the HFTBA transverse plane, a significant difference was observed at 9-22% ($p = 0.043$). A significant difference was also observed in knee transverse plane kinematics at 59-72% ($p = 0.039$). Lastly, a significant difference was also seen in the trunk transverse plane at 92-100% ($p = 0.048$). No other significant differences were observed between kinematic variables during this phase of cutting (Figure 5.11 and Figure 5.12).

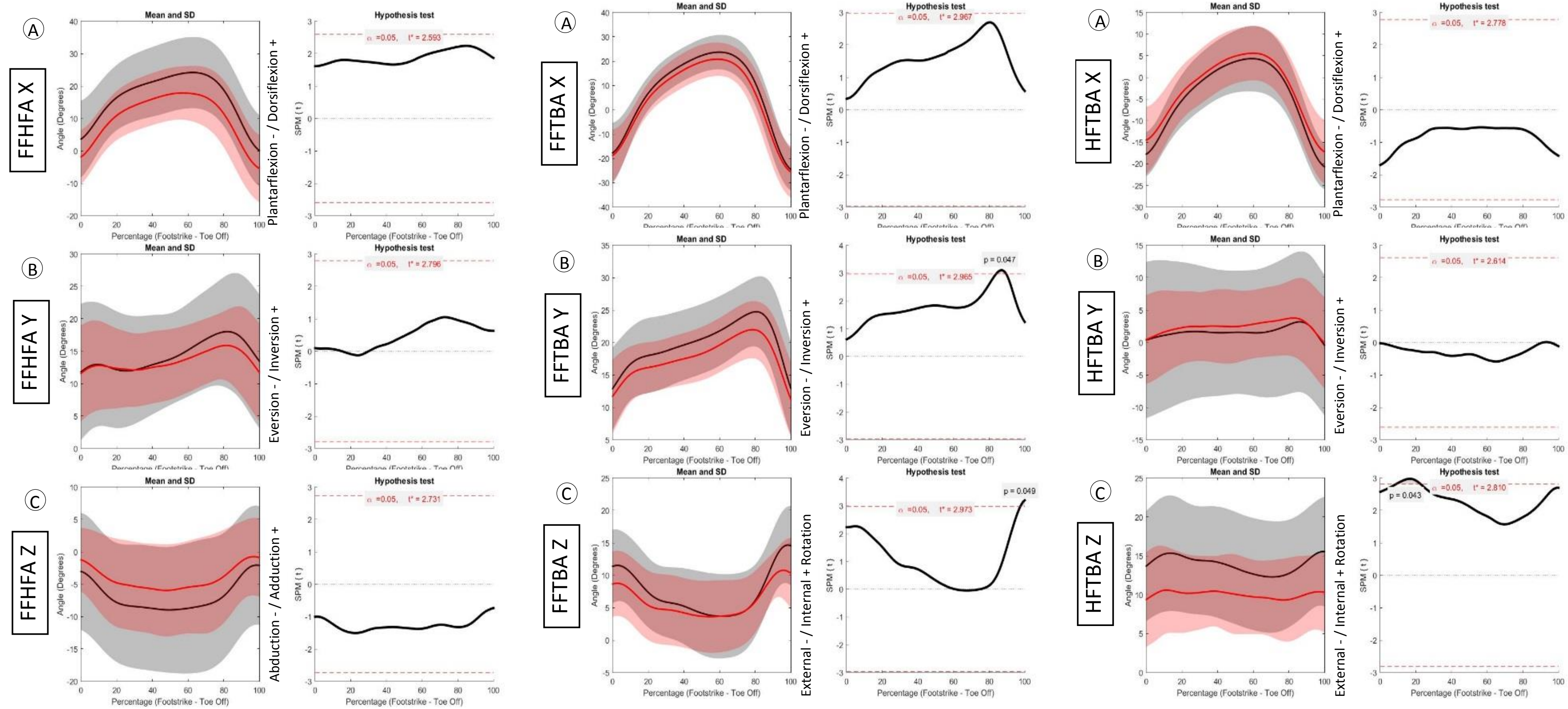


Figure 5.11 Cutting - FFHFA, FFTBA and HFTBA displacement (x, y, z) Foot strike to toe off - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

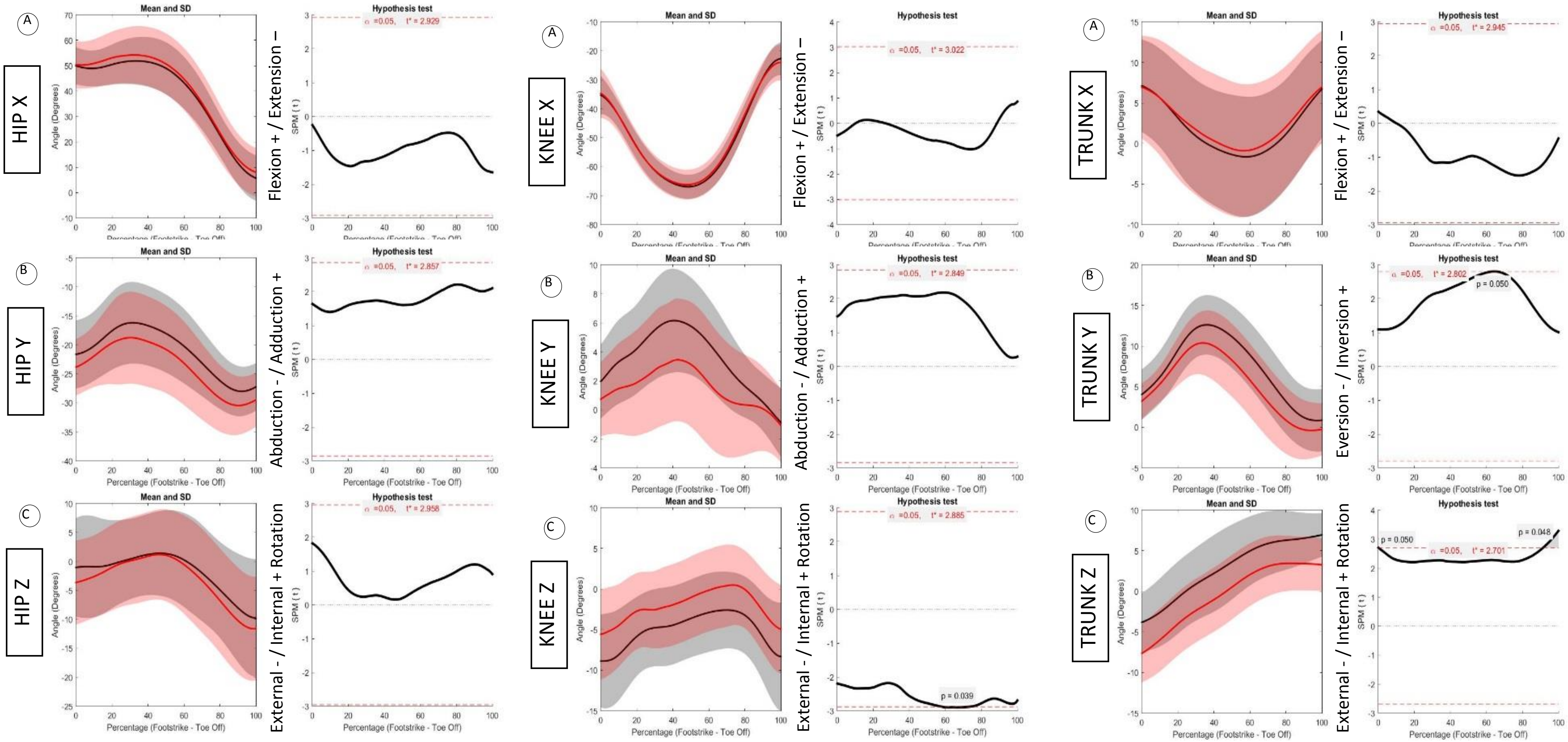


Figure 5.12 Cutting - Hip, Knee and Trunk displacement (x, y, z) Foot strike to toe off - mean ± SD (Dominant - — Non-Dominant - —) and *t*-test output.

5.1.4 Discussion

This preliminary study aimed to identify whether differences existed in kinematics between the dominant and the non-dominant limb. These findings are summarised in Table 5.1.

Table 5.1 Summary of findings of SPM analysis between the dominant and the non-dominant limb

		Findings when comparing dominant to non-dominant limb
Walk	Heel strike to toe off	Significant differences were observed in: • HFTBA transverse plane at 83-88% (*p* = 0.049) • Hip transverse plane at 0-36% (*p* = 0.014) • Knee transverse plane at 0-14% (*p* = 0.042) and 20-99% (*p* = <0.001) • Trunk transverse plane at 0-100% (*p* = 0.008)
	Toe off to heel strike	Significant differences were observed in: • HFTBA transverse plane at 89-96% (p = 0.048) • Knee transverse plane at 34-46% (p = 0.044) and 61-100% (p = 0.015) • Trunk transverse plane at 0-100% (p = 0.023)
Single Leg Land	200 ms prior to initial contact to initial contact	Significant differences were observed in: • Hip frontal plane at 0-11% (p = 0.049) • Knee transverse motion from 0-58% (p = 0.011)
	Initial contact to 200 ms post initial contact	Significant differences were observed in: • Knee transverse plane at 0-20% (p = 0.046) and 76-100% (p = 0.045) • Trunk frontal plane at 82-100% (p = 0.046)
Cut	200 ms prior to initial contact to foot strike	Significant differences were observed in: • FFTBA transverse plane at 46-82% (p = 0.014) • HFTBA transverse plane at 22-77% (p = 0.032) • Trunk transverse plane at 0-100% (p = 0.014)
	Foot strike to toe-off	Significant differences were observed in: • FFTBA frontal plane at 84-89% (p = 0.047) • FFTBA transverse plane at 98-100% (p = 0.049) • HFTBA transverse plane at 9-22% (p = 0.043) • Knee transverse plane at 59-72% (p = 0.039) • Trunk transverse plane at 92-100% (p = 0.048)

As previously identified, lateral ankle sprains are not exclusive to either the dominant or the non-dominant limb (Woods et al., 2003; Yeung et al., 1994). Clinically, it is crucial to identify differences in kinematics between individuals with ankle instability and healthy controls regardless of whether the affected side is the dominant or non-dominant. This study identified a handful of differences that are present when comparing the dominant to the non-dominant limb in healthy participants.

Clinically, comparison is drawn to the non-injured side to determine severity and criteria for return to play. When comparing biomechanics between healthy controls and those with lateral instability some discuss dominance for the control group but neglect to report this within the ankle instability group (Knight & Weimar, 2011b). Others match the side and ignore the involvement of dominance (Kipp & Palmieri-Smith, 2013). This preliminary study has highlighted that differences are present when comparing the dominant and non-dominant limbs. Throughout this thesis, the control group will be matched for dominance to the instability group (if the dominant limb is the affected limb the matched control will also be the dominant limb) to ensure that the dominance effect is accounted for within the analysis.

Chapter 6.0 Study 1 – Analysis of Walking Muscle Activation and Kinematics in Individuals with Ankle Instability and Healthy Control Participants

6.1 Walking Chapter Overview

Walking is of high importance in daily life and is often problematic for people with CAI who complain of giving way sensations on uneven and level surfaces (Wright et al., 2013a). Research suggests that the position of the affected ankle joint at specific time points during the gait cycle may predispose an ankle to injury, this may be associated with or caused by ankle joint instability (Delahunt et al., 2006a). Research analysing frontal plane ankle kinematics during walking observed increased ankle inversion that corresponded to greater ankle inversion during more sport-specific tasks such as jump-landing (Donovan & Feger, 2017). Gait analysis is often used in the development of rehabilitation and injury prevention protocols. Therefore changes in full body gait kinematics must be investigated, and where possible accounted for, as these may impact not only walking but other more dynamic movements.

Previous literature investigating CAI during walking has modelled the foot as one rigid segment (Monaghan et al., 2006; Stebbins et al., 2006). Rigid segment modelling excludes motion between different segments of the foot providing inadequate information on the biomechanics of the foot (Stebbins et al., 2006). De Ridder et al. (2013) appears to be the first study to analyse walking using a multi-segmental foot model, comparing the use of the Ghent Foot Model to a rigid foot model in participants with CAI, copers (no symptoms of instability after a recent ankle sprain) and control participants. They concluded that the multi-segmental foot model provided greater details of the intricacies of the foot, showing differences between segments when comparing groups.

Potential relationships between proximal adaptations and injury have also been noted (Doherty et al., 2016a), therefore full body kinematic analysis is warranted. Upper body kinematic analysis should be considered when investigating changes in the lower extremities as there may be a significant relationship with changes observed in proximal segments (Doherty et al., 2016a). The body is a multi-linked system with the rectus femoris, hamstrings, and gastrocnemius muscles crossing the hip, knee and ankles. The kinetic chain concept suggests that movement of the trunk during landing (which accounts for 35.5% body mass) will also have an impact on motion of the hip and therefore knee and ankle (Kulas et al., 2008).

It is pertinent to gain full knowledge of kinematic movement patterns, prior to implementing intervention strategies, to ensure these will not subsequently impact other joints within the kinetic chain. To the author's knowledge, no research has combined trunk kinematics with a full lower limb and multi-segmental foot model to address, in combination, the possible proximal and distal differences between groups. This is a 3 part study to compare walking kinematics and muscle activation patterns between individuals with CAI and healthy controls. Chapter 6.2 uses full time series analysis to identify differences in movement and muscle activation patterns. Chapter 6.3 addresses the impact of angular displacement, angular velocities and angular accelerations to observe whether the speed of angular movement is an impacting factor in ankle instability during walking. The final part of this study (Chapter 6.4) will address whether the variables where significant differences were identified in Chapter 6.3 can be used to predict the score of individuals on the IdFAI questionnaire.

6.2 Study 1 – Part 1 - Full Gait Cycle Analysis of Lower Limb and Trunk Kinematics and Muscle Activations During Walking in Participants with And Without Ankle Instability

Published in Gait & Posture (July 2018)

Northeast, L., Gautrey, C. N., Bottoms, L., Hughes, G., Mitchell, A. C., & Greenhalgh, A. (2018). Full gait cycle analysis of lower limb and trunk kinematics and muscle activations during walking in participants with and without ankle instability. *Gait & Posture*, *64*, 114-118. https://doi.org/10.1016/j.gaitpost.2018.06.001.

Conference Communication

Northeast, L., Gautrey, C., Mitchell, A., Bottoms, L. & Greenhalgh, A. (2018). A comparison of lower limb kinematics and electromyography during walking between athletes with chronic ankle instability and healthy controls. World Congress of Biomechanics, Dublin, Ireland, 8-12 July 2018.

6.2.1 Introduction

Prior research reports joint angles and muscle activation characteristics at discrete time points during walking (Koldenhoven et al., 2016; Monaghan et al., 2006), rather than whole kinematic time-series curves. Biomechanical data is one dimensional (1D) (time and kinematic or force trajectories), therefore this may result in focus bias or missing potential significance or trends during other phases of the gait cycle (Pataky et al., 2013). Statistical parametric mapping is a concept introduced to biomechanics from brain research (Friston et al., 1994) which enables curve analysis across the whole movement (Pataky et al., 2013). Comparison between SPM and time series analysis using confidence intervals concluded SPM to be the most suitable method for analysis of 1D data, due to increased generalisability of probabilistic conclusions (with the use of hypothesis testing techniques) and the ability to present results in a more consistent manner aiding interpretation of findings (Pataky et al., 2015). De Ridder et al. (2013) used SPM to compare foot kinematics between participants with CAI, copers and controls, identifying exact time periods of significantly increased forefoot inversion within the stance phase of walking in the CAI group and the copers group when compared to the control.

Previous literature investigating sEMG found hip abductor weakness to be associated with acute ankle sprains, though it is unclear whether this is a cause or an effect of the sprain (Friel et al., 2006). Koldenhoven et al. (2016) reported increased gluteus medius activation in the late stance and early swing phase of walking in CAI participants, suggesting this may be a coping mechanism used to generate a wider base of support or to increase lower limb stability. Decreased tibialis anterior activation was also observed, resulting in increased ankle plantarflexion prior to HS, though the possible reasoning for this was not suggested. This loose-packed position (joint capsule lax and joint surfaces are not congruent) has been found to be unstable (Hopkins et al., 2012), putting the individual at an increased risk of ankle sprains.

It is suggested that combined analysis of the trunk, hip, knee and multi-segmental foot kinematics and sEMG activation patterns across the stance and swing phases of gait will provide greater insight into possible differences that exist, not just within the foot, but across the full kinetic chain. This may provide greater insight to clinicians rehabilitating those with ankle instability and may highlight areas of importance in the reduction of future ankle sprains. The aim of this study was to compare trunk, hip, knee and multi-segmental foot kinematics and muscle activation during the stance and swing phase of walking between participants with CAI and healthy controls.

This study will address the following hypotheses previously presented in Chapter 2.19:

- **H1** - CAI participants will display modified kinematic movement patterns (SPM) during walking
- **H2** - CAI participants will display modified muscle activation patterns (SPM) during walking

6.2.2 Methods

Participants

Eighteen healthy controls (age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 participants with CAI (age: $\bar{\chi}$ = 22.0 ± 2.7 years, height: $\bar{\chi}$ = 176.8 ± 7.9 cm, mass: $\bar{\chi}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were included in this study as outlined in General Methods 3.1. Dominance was determined by asking which leg they would use to kick a ball in line with previous research (Hopkins et al., 2012; Wright et al., 2013b).

Protocol

Participants completed a 5-minute warm-up on a cycle ergometer (Monark Ergomedic 874E, Sweden) at 60 Watts. Electromyographic data were recorded as outlined in General Methods 3.4, bilaterally for the gluteus medius and tibialis anterior. Maximum voluntary isometric contractions were performed 3 times for a 5-second duration. Peak activation of three trials was identified as the MVIC which was used to allow comparison between participants' sEMG data. Gluteus medius MVIC was performed in side-lying with the participant maximally abducting their hip (positioned mid-range) into a rigid strap positioned proximal to the knee (Hislop & Montgomery, 2007). Tibialis anterior MVIC was performed in a seated position and the participant maximally dorsiflexing and inverting their foot against a rigid strap across the forefoot (Hislop & Montgomery, 2007). Voluntary contractions were used to inspect recordings for crosstalk.

Full body and multi-segmental foot model 3D kinematics were recorded as outlined in General Methods 3.5. Participants were instructed to walk at their normal walking speed through the calibrated capture volume. Pace was not controlled, as this was deemed to be unnatural and has been previously shown to impact on stride time variability due to increased central nervous system involvement (Springer & Gottlieb, 2017). Participants walked barefoot 3.5 m before data were collected (Najafi et al., 2010) and proceeded for 7 m across the walkway. Walking speed was recorded using pelvis segment velocity. Barefoot walking was used in accordance with the method of De Ridder et al. (2013) and due to the number of markers on the foot. Participants performed a familiarisation until they were comfortable with the movement, before recording three trials for analysis (Mullineaux, Bartlett, & Bennett, 2001). Trials were deemed successful when all tracking markers were in view of the cameras (observed on screen).

Data and Statistical Analysis

Data were inspected using Cortex software (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) before importing into Visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland). Data were smoothed using a 6 Hz Butterworth filter. Initial contact was determined using the method proposed by O'Connor et al. (2007), which creates a new signal by calculating the midpoint between the posterior inferior heel marker and the toe marker (between 2nd and 3rd metatarsal heads). The first derivative was calculated on the vertical component of the signal. Event markers were created at the minimum value for HS and maximum value for toe-off. Electromyographic data were root mean squared by a moving window of 100 ms and normalised to MVIC. Visual inspection of the data identified noise in the signal for two of the participants, warranting their sEMG data be removed. To maintain pre-experimental research design, matched controls assigned to the two participants also had their sEMG data removed. Kinematic and sEMG data were exported for the stance (HS to toe off) and swing (toe-off to HS) phases into MATLAB R2015a (The Math Works, Natick, Massachusetts) to perform SPM analysis.

Kinematic data were exported for forefoot-hindfoot angle, forefoot-tibia angle, hindfoot-tibia angle, hip, knee and trunk angles in the sagittal, frontal and transverse planes of motion. So not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment (De Ridder et al., 2013; Wright et al., 2011). Data were analysed using SPM in MATLAB (SPM1D open-source package, spm1d.org). Normality was tested using the D'Agostino-Pearson's test. A matched control limb was compared to the CAI groups' affected limb using an independent-samples *t*-test ($\alpha = 0.05$). The unaffected and affected limb of the CAI group were compared using a paired-samples *t*-test ($\alpha = 0.05$). A matched control limb was compared to CAI groups' unaffected limb using an independent-samples *t*-test ($\alpha = 0.05$).

6.2.3 Results

Participant characteristics

Independent-samples *t*-tests revealed no significant differences ($p > 0.05$) between groups for age, height, mass, or shoe size. An independent-samples *t*-test reported no significant difference in walking velocity when comparing the control group (1.20 ± 0.15 $m.s^{-1}$), and CAI group (1.18 ± 0.09 $m.s^{-1}$).

Gait analysis

No significant differences were observed in FFHFA (Appendix E, Figure E.1, Figure E.8), FFTBA (Appendix E, Figure E.2, Figure E.9), HFTBA (Appendix E, Figure E.3, Figure E.10), hip (Appendix E, Figure E.4, Figure E.11), knee (Appendix E, Figure E.5, Figure E.12), or trunk (Appendix E, Figure E.6, Figure E.13) angles in the sagittal, frontal, or transverse planes of motion, in the stance or swing phase, between the matched control and the CAI groups' affected limb. No significant differences were observed in the gluteus medius or tibialis anterior muscle activation (Appendix E, Figure E.7, Figure E.14) in either phase of gait between the matched control and the CAI groups affected limb.

A significant difference was reported between the CAI groups' unaffected and affected limb in the FFTBA in the frontal plane, where increased inversion was observed in the affected limb at 4-16% of the stance phase (mean difference = 3.07°, peak difference = 3.24°, p = 0.039, Figure 6.1).

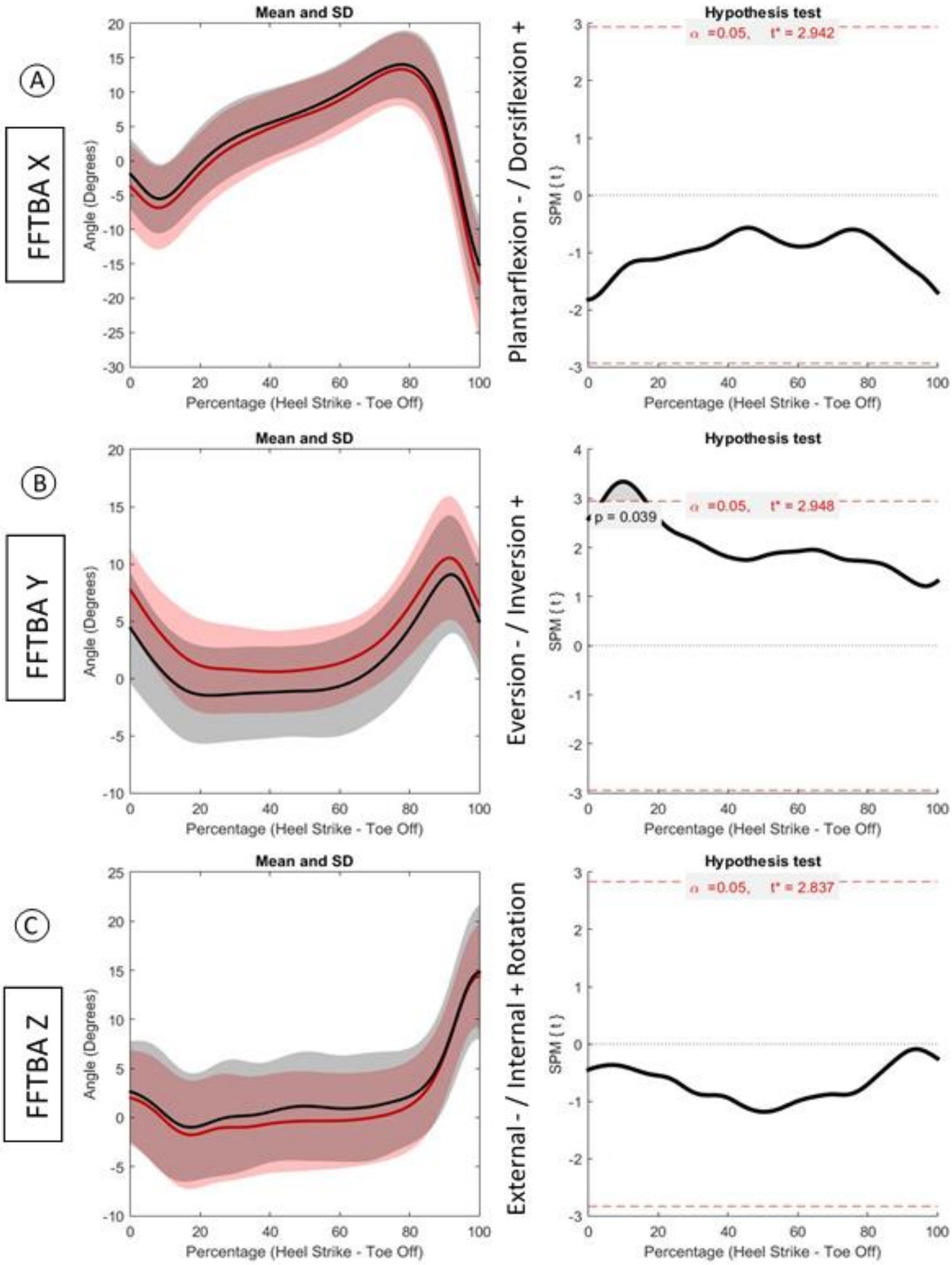


Figure 6.1 Forefoot – Tibia angles (x, y, z) Heel strike - Toe off – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

No other significant differences were reported for FFTBA during HS to toe off (Figure 6.1) or toe off to HS (Appendix E, Figure E.22). Furthermore, no significant differences were noted between FFHFA (Appendix E, Figure E.15, Figure E.21), HFTBA (Appendix E, Figure E.16, Figure E.23), hip (Appendix E, Figure E.17, Figure E.24), knee (Appendix E, Figure

E.18, Figure E.25), or trunk (Appendix E, Figure E.19, Figure E.26) angles or in muscle activation of the tibialis anterior and gluteus medius (Appendix E, Figure E.20, Figure E.27) between the unaffected and affected limbs at any time point. Finally, no significant differences were observed between the CAI groups' unaffected limb and the control groups' limb (matched for dominance) in any of the recorded variables in either the stance or swing phases of movement (Appendix E, Figure E.28-Figure E.41).

6.2.4 Discussion

The aims of this study were to explore the differences in kinematics and muscle activation patterns between CAI participants' unaffected and affected ankles and to compare these to a matched control group throughout the gait cycle. The results of this study are comparable to a readily available published data set (Fukuchi, Fukuchi, & Duarte, 2018).

Increased FFTBA inversion was found in the affected limb of the CAI group when compared to its unaffected counterpart at 4-16% stance. As differences were observed in kinematics null hypothesis for $H0_1$ can be rejected. Total range of motion in the frontal plane has previously been reported to be 35 degrees (Brockett & Chapman, 2016) thus a mean difference of 3.07° and peak difference of 3.24° may be a clinically significant finding. This finding supports previous hypotheses that participants with CAI may exhibit altered joint position sense and proprioceptive awareness (Konradsen, 2002). Increased inversion at ground contact decreases bony restrictions of the foot-ankle complex, thus, when loaded with bodyweight, increases inversion torque and joint susceptibility to injury (Konradsen, 2002). The early period of the stance phase is not consciously mediated (Lees, 1981), thus increased inversion places the ankle in a position of increased vulnerability at HS, potentially predisposing the affected limb to further ankle sprains and episodes of giving way. Whilst not within the remit of this study, differences in angular displacement associated with CAI may be exacerbated during more dynamic movements e.g. cutting, single/double leg landing, running, or when walking on uneven surfaces, as research has previously shown increased kinematics in walking often correspond to increased kinematics during more dynamic sporting activities (Donovan & Feger, 2017).

The lack of significant differences at the hip or knee, between groups, in the frontal, sagittal or transverse planes of motion in the current study is consistent with the findings of Monaghan et al. (2006), who found no significant differences in hip and knee kinematics between participants with CAI and healthy control participants from 100 ms pre-HS to 200 ms post-HS. Within the current study, trunk kinematics were measured in all three planes, however, no significant differences were identified between groups suggesting that no proximal adaptations took place within the CAI group during walking.

No significant differences were observed in tibialis anterior or gluteus medius muscle activation between groups during gait meaning that null hypothesis $H0_2$ can be accepted. This is contrary to the findings of Hopkins et al. (2012) who when reporting discrete peak value data, observed an increase in tibialis anterior activation from 15-30% and 45-70% of stance, which they speculated was a motor strategy to maintain a more dorsiflexed, stable position in the affected limb compared to a dominance matched control limb. Methodological differences exist between the current study and the study by Hopkins et al. (2012) as participants walked shod rather than barefoot as in the present study which may have caused adaptation to occur. Decreased muscle activation patterns have previously been observed in barefoot walking compared to shod walking (Franklin, Grey, Heneghan, Bowen, & Li, 2015). Hopkins et al. (2012) also examined tibialis anterior activation whilst walking on a treadmill rather than over ground. These differences in methodological approaches may account for the differing results between the two studies. Koldenhoven et al. (2016) recorded significantly higher gluteus medius muscle activation in the final 50% of stance and the first 25% of the swing phase when compared to healthy participants, however, this was again performed shod on a treadmill, making comparisons with the current study difficult. Previous studies have found differing muscle activation patterns and sagittal plane motion with treadmill walking compared to overground walking (S. J. Lee & Hidler, 2008). Therefore, the results of this study may prove a more valid representation of the everyday task of overground walking. Furthermore, comparison to previous research may not be appropriate due to the different statistical analysis used (Franklin et al., 2015; Hopkins et al., 2012; Koldenhoven et al., 2016; S. J. Lee & Hidler, 2008). It is important to note that grouping of participants was purely through the inclusion criteria outlined in the IAC guidelines and with use of the IdFAI questionnaire (Gribble et al., 2013) and no other discriminative measures e.g. Beighton score for hypermobility were used. This may be a limitation although further research is required to establish this. It is important to note the high variability that can be observed within the EMG data particularly in the CAI group data. This may be due to the difficulties with obtaining true MVIC's particularly with clinical populations (Konrad, 2005) as was the case with this cohort of participants. This may also be explained by the change in muscle lengths that occur during dynamic movements, motor unit synchronisation and increased superposition of electrical activity during dynamic movements (Konrad, 2005).

This study observed no differences in gait biomechanics between healthy controls and participants with CAI, however, differences were found between affected and unaffected limbs of the CAI group. This may suggest greater inversion during the stance phase is a direct result of the ankle sprain or a predisposing factor for injury. This may also support ideas in the literature that some individuals are biomechanically predisposed to CAI and why the incidence of bilateral CAI are so high. Early gait re-education could be warranted as individuals return to walking to avoid the development of compensatory strategies. This statement is made with caution as a prospective study is warranted to truly determine whether greater inversion is present prior to or as a result of the injury.

This study analysed kinematic and electromyographic parameters to determine differences in movement patterns and muscle activations. Future research should identify the impact of CAI on kinetic parameters using full curve analysis to identify differences between groups. Further research should use these analysis methods to examine dynamic movements such as change of direction, single and double leg landing and running gait. Analysis of additional muscle sEMG signals may also provide a greater understanding of potential differences between groups. In particular muscles such as the peroneals which may be a causative factor of the differences observed in FFTBA frontal plane kinematics.

6.2.5 Conclusion

Participants with CAI exhibited increased inversion patterns during the stance phase of gait in their affected limb compared to their unaffected limb as the available range in the frontal plane is only 35 degrees a peak difference of 3.24 degrees warrants further investigation and may prove a key area of focus for future investigations. This change in movement pattern may predispose those with CAI to repeated episodes of giving way and further ankle sprains. Increased inversion may also be a significant risk factor in more dynamic movements, thus further research should investigate these using a multi-segmental foot model. Incorporating kinetic variables into this analysis may also be beneficial to determine differences in ground reaction forces and moments.

6.2.6 Development of Research Within the Thesis

Statistical parametric mapping allowed in-depth analysis of the whole pattern of movement during walking. Excluding the one significant difference (increased FFTBA inversion in the affected limb of the CAI group at 6-16% stance) the patterns of movement are similar between groups. This method, however, time normalises the data to enable full curve analysis between groups and as such further discrete analysis of the data set investigating peak angular displacements, peak angular velocities and peak angular accelerations is warranted in order to highlight further differences if present between these groups.

6.3 Study 1 – Part 2 - A Comparison of Lower Limb Angular Displacements, Angular Velocities and Angular Accelerations During Walking Between Participants with Chronic Ankle Instability and Healthy Controls

Conference Communication

Northeast, L., Gautrey, C., Mitchell, A., Bottoms, L. & Greenhalgh, A. (2018). A comparison of lower limb angular displacements, velocities and accelerations during walking between athletes with chronic ankle instability and healthy controls. World Congress of Biomechanics, Dublin, Ireland, 8-12 July 2018.

6.3.1 Introduction

Research investigating movement patterns during gait only observed significantly increased forefoot-tibia (FFTBA) inversion angular displacement (4-16% stance) in the CAI group's affected limb when compared to the unaffected (Northeast et al., 2018). Tilt platform research observed that affected ankles cover the same angular displacement in a significantly shorter duration (Vaes, Van Gheluwe, & Duquet, 2001), suggesting angular velocity may be a risk factor for CAI. Potentially damaging increases in ankle inversion velocity have been reported during walking at 100 ms pre-HS to 200 ms post-HS (Monaghan et al., 2006). Data reported from an accidental LAS sustained during cutting have reported increased plantarflexion, internal rotation and inversion angular velocities at the ankle joint (Fong et al., 2009a). These findings suggest that angular velocity may be crucial in differentiating movement characteristics of participants with and without CAI.

The aim of this study was to compare angular displacement, angular velocity and angular acceleration of the trunk, hip, knee and foot during gait at HS, 100 ms pre-HS to HS, and HS to 200 ms post-HS, between participants with CAI and healthy controls to determine whether an affected limb demonstrates different kinematics, whether a potential compensatory strategy is adopted between limbs and whether the ‘unaffected’ limb is comparable to a true control limb.

This study will address the following hypothesis previously presented in Chapter 2.19:

- **H$_3$** - CAI participants will display modified discrete kinematic variables during walking

6.3.2 Method

Participants

Eighteen healthy controls (14 males, 4 females, age: $\bar{x}$ = 22.4 ± 3.6 years, height: $\bar{x}$ = 177.8 ± 7.6 cm, mass: $\bar{x}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 participants with CAI (13 males, 5 females, age: $\bar{x}$ = 22.0 ± 2.7 years, height: $\bar{x}$ = 176.8 ± 7.9 cm, mass: $\bar{x}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were included in this study as outlined in General Methods 3.1.

Protocol

Kinematic data from the walking trials collected in Chapter 6.2 were inspected using Cortex software (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) and imported into visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland, USA) for analysis within this study.

Data Reduction and Statistical Analysis

Data were smoothed using a 6 Hz Butterworth filter and HS was determined using the method proposed by O’Connor et al. (2007). Event markers were created at the minimum value for HS, 100 ms prior to HS and 200 ms post HS. Kinematic data were exported for minimum and maximum angular displacement, angular velocity and angular acceleration for 100 ms pre-HS to HS, at HS and HS to 200 ms post-HS. Kinematic data were exported for the forefoot-hindfoot angle, forefoot-tibia angle, hindfoot-tibia angle, hip, knee and trunk angles in all planes of motion. So not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment (De Ridder et al., 2013; Wright et al., 2011). Statistical analysis was performed in SPSS (IBM SPSS Statistics for Windows, Version 24.0. Armonk, NY: IBM Corp). The matched control limb was compared to the CAI group's affected limb and unaffected limb using an independent samples t-test ($\alpha = 0.05$). The unaffected and the affected limb of the CAI group were compared using a paired samples t-test ($\alpha = 0.05$). Data is presented as group means and standard deviations and the symbol Δ denotes differences between group means.

6.3.3 Results

Independent samples t-tests revealed no significant differences ($p > 0.05$) between groups for age, height, mass, or shoe size. An independent samples t-test reported no significant difference in walking velocity between the control group (1.20 ± 0.15 m/s), and CAI group (1.18 ± 0.09 m/s).

CAI Group Affected Limb - Matched Control

100 ms pre-HS to HS - Significantly decreased FFTBA peak internal rotation ($\Delta = 4.95°$, $p = 0.002$) and external rotation ($\Delta = 4.77°$, $p = 0.002$) angular displacement and a significantly decreased peak trunk lateral flexion angular displacement towards the stepping (affected limb) ($\Delta = 1.67°$, $p = 0.007$) were observed in the affected limb (Table 6.1). Significantly decreased FFTBA dorsiflexion angular velocity ($\Delta = 12.19°/s$, $p = 0.030$) and significantly increased peak positive FFTBA transverse plane angular acceleration (Δ

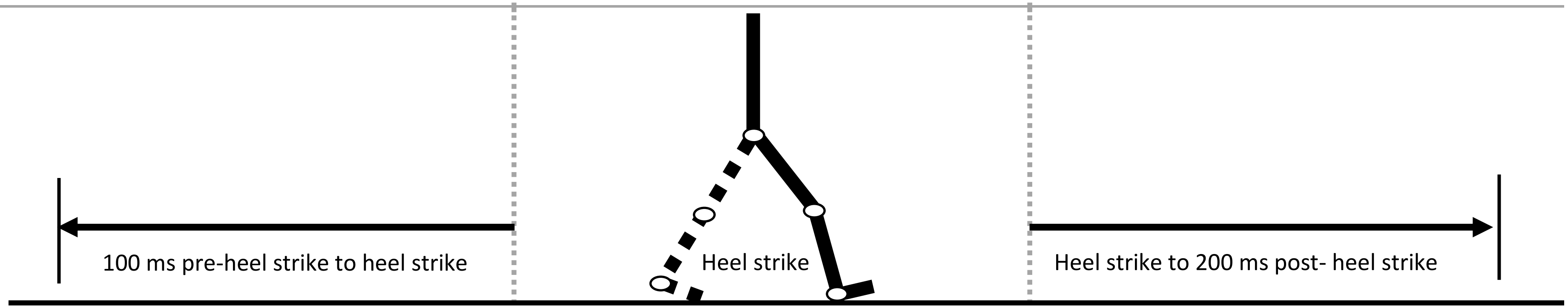
= 305.11°/s^2, p = 0.037) and decreased peak negative knee frontal plane angular acceleration (Δ = 257.12°/s^2, p = 0.040) were also seen in the affected limb (Table 6.1). No other significant differences were observed at this time point (Appendix F, Table F.1, Table F.4, Table F.7).

Heel strike - Significantly decreased FFHFA abduction (Δ = 21.23°/s, p = 0.012), significantly decreased FFTBA external rotation (Δ = 25.90°/s, p = 0.007) and significantly increased knee adduction (Δ = 11.20°/s, p = 0.014) angular velocity were observed in the CAI affected limb as well as significantly decreased FFTBA negative frontal plane angular acceleration (Δ = 815.55°/s^2, p = 0.006) (Table 6.1). No significant differences were observed in angular displacements between groups (Appendix F, Table F.2). No other significant differences were observed in angular velocity or angular acceleration at this time point (Appendix F, Table F.5, Table F.8).

HS to 200 ms post-HS - Significantly decreased peak FFTBA internal rotation angular displacement (Δ = 4.55°, p = 0.007) and significantly decreased peak trunk lateral flexion angular displacement towards the stepping limb (affected limb) (Δ = 1.20°/s, p = 0.042) was observed in the affected limb of the CAI group along with significantly decreased FFTBA external rotation angular velocity (Δ = 20.45°/s, p = 0.007). A significantly decreased peak negative frontal plane trunk acceleration (Δ = 331.55°/s^2, p = 0.003) and significantly increased peak negative transverse plane trunk acceleration (Δ = 142.92°/s^2, p = 0.008) was observed in the affected limb (Table 6.1). No other significant differences were observed at this time point (Appendix F, Table F.3, Table F.6, Table F.9).

Table 6.1 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the affected limb of the CAI group to a matched control.

ANGULAR DISPLACEMENT	- ↓ Forefoot-Tibia Internal Rotation (affected: 2.30 ± 4.33, control: 7.25 ± 4.37, *p* = 0.002) - ↓ Forefoot-Tibia External Rotation (affected: 0.60 ± 4.18, control: 5.37 ± 4.22, *p* = 0.002) - ↓ Trunk lateral flexion towards the stepping (affected) limb (affected: 0.05 ± 1.87, control: -1.62 ± 1.64, *p* = 0.007)	- No significant differences observed	- ↓ Forefoot-Tibia Internal Rotation (affected: 1.72 ± 4.72, control: 6.27 ± 4.74, *p* = 0.007) - ↓ Trunk lateral flexion towards the stepping (affected) limb (affected: 0.59 ± 1.82, control: -0.61 ± 1.59, *p* = 0.042)
ANGULAR VELOCITY	- ↓ Forefoot-Tibia Dorsiflexion (affected: -2.20 ± 13.24, control: 9.99 ±18.60, *p* = 0.030)	- ↓ Forefoot-Hindfoot Abduction (affected: -6.39 ± 28.21, control: -27.62 ± 19.21, *p* = 0.012) - ↓ Forefoot-Tibia External Rotation (affected: -9.71 ± 28.78, control: -35.61 ± 25.53, *p* = 0.007) - ↑ Knee Adduction (affected: 6.95 ± 12.09, control: -4.25 ± 13.76, *p* = 0.014)	- ↓ Forefoot-Tibia External Rotation (affected: -60.41 ± 20.60, control: -80.86 ± 21.96, *p* = 0.007)
ANGULAR ACCELERATION	- ↑ Forefoot-Tibia transverse peak +ve (affected: 906.03 ± 400.22, control: 600.92 ± 443.11, *p* = 0.037) - ↓ Knee frontal peak -ve (affected: -326.44 ± 339.92, control: -583.56 ± 381.81, *p* = 0.040)	- ↓ Forefoot-Tibia Eversion (affected: -883.42 ± 770.10, control: -1698.97 ± 908.53, *p* = 0.006)	- ↓ Trunk frontal peak –ve (affected: -554.51 ± 224.99, control: -886.06 ± 386.39, *p* = 0.001) - ↑ Trunk transverse peak –ve (affected: -479.51 ± 149.42, control: -336.59 ± 157.01, *p* = 0.003)



CAI Group Unaffected Limb - Matched Control

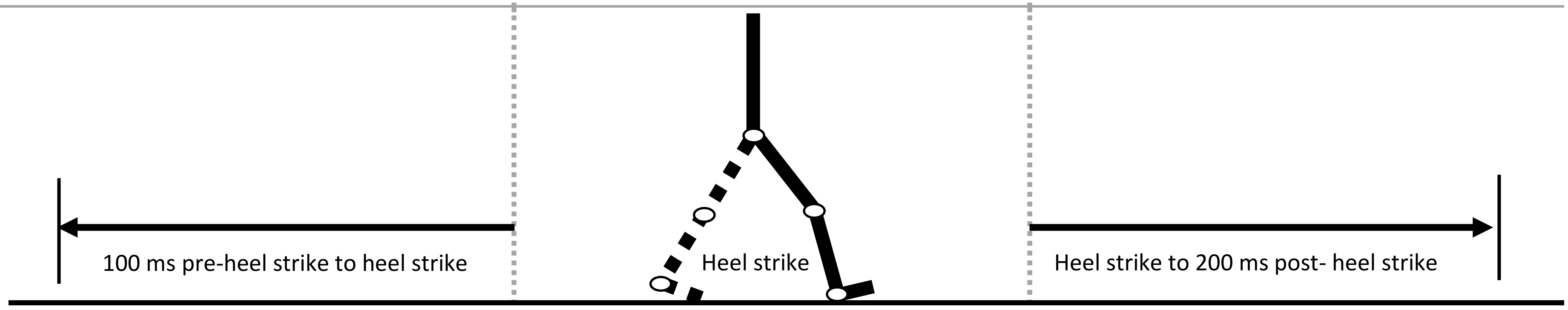
100 ms pre-HS to HS - Significantly decreased trunk lateral flexion angular displacement away from the stepping limb (towards the affected limb) (Δ = 1.17°, p = 0.037) and significantly decreased FFTBA external rotation velocity (Δ = 73.53°/s, p = 0.047) were observed in the CAI group unaffected limb (Table 6.2). No other significant differences were observed at this time point (Appendix F, Table F.1, Table F.4, Table F.7).

Heel strike - Significantly decreased FFHFA inversion angular displacement (Δ = 5.17°/s, p = 0.039), significantly decreased FFTBA internal rotation angular displacement (Δ = 3.41°/s, p = 0.045) and significantly increased knee adduction angular velocity (Δ = 8.46°/s^2, p = 0.037) were also observed in the unaffected limb (Table 6.2). No other significant differences were observed between the unaffected limb and the matched control at this time point (Appendix F, Table F.2, Table F.5, Table F.8).

HS to 200 ms post-HS - A significantly decreased trunk lateral flexion angular displacement away from the stepping limb (towards the affected limb) (Δ = 1.51°, p = 0.028) and significantly decreased peak FFHFA eversion angular velocity (Δ = 19.76°/s, p = 0.033) were observed (Table 6.2). No other significant differences were observed between groups from HS to 200 ms post HS (Appendix F, Table F.3, Table F.6, Table F.9).

Table 6.2 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the unaffected limb of the CAI group to a matched control.

ANGULAR DISPLACEMENT	- ↓ Trunk lateral flexion away from the stepping limb (towards the affected limb) (unaffected: -0.51 ± 1.87, control: 0.66 ± 1.33, p = 0.007)	- ↓ Forefoot-Hindfoot Inversion (unaffected: 3.43 ± 7.37, control: 8.60 ± 7.07, p = 0.039) - ↓ Forefoot-Tibia Internal Rotation (unaffected: 2.75 ± 5.17, control: 6.16 ± 4.64, p = 0.045)	- ↓ Trunk lateral flexion away from the stepping limb (towards the affected limb) (unaffected: 4.22 ± 2.27, control: 5.73 ± 1.60, p = 0.028)
ANGULAR VELOCITY	- ↓ Forefoot-Tibia External Rotation (unaffected: -29.90 ± 20.93, control: -43.63 ± 18.97, p = 0.047)	- ↓ Knee Adduction (unaffected: 3.73 ± 11.94, control: -4.73 ± 11.40, p = 0.037)	- ↓ Forefoot-Hindfoot Eversion (unaffected: -53.52 ± 21.72, control: -73.28 ± 30.76, p = 0.033)
ANGULAR ACCELERATION	- No significant differences observed	- No significant differences observed	- No significant differences observed



CAI Group Affected Limb - CAI Group Unaffected Limb

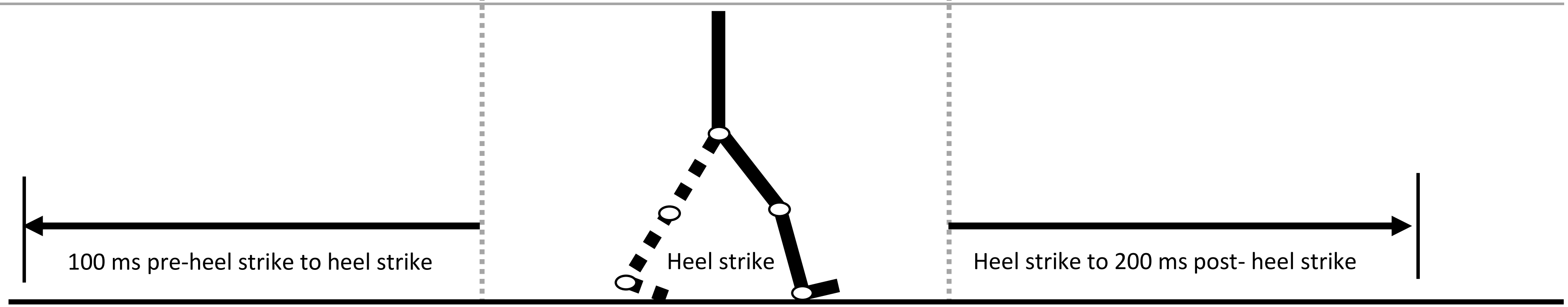
100 ms pre-HS to HS - No significant differences were observed pre-HS (Appendix F, Table F.1, Table F.4, Table F.7).

Heel strike - Significantly increased FFTBA inversion angular displacement (Δ = 2.61˚, *p* = 0.041) was observed in the affected limb when compared to the unaffected limb (Table 6.3). No other significant differences were observed at this time point (Appendix F, Table F.2, Table F.5, Table F.8).

HS to 200 ms post-HS - Significantly increased peak FFTBA inversion angular displacement (Δ = 2.45˚, *p* = 0.043) and significantly decreased peak FFTBA eversion angular displacement (Δ = 2.25˚, *p* = 0.018) were observed in the affected limb, along with significantly decreased peak FFTBA inversion velocity (Δ = 6.57˚/s, *p* = 0.031) and significantly increased peak trunk external rotation velocity (Δ = 3.03˚/s, *p* = 0.015) (Table 6.3). Significantly decreased peak positive FFTBA transverse plane acceleration (Δ = 303.26˚/s^2, *p* = 0.014), significantly increased peak negative sagittal plane trunk acceleration (Δ = 50.30˚/s^2, *p* = 0.042), significantly decreased peak negative trunk frontal plane acceleration (Δ = 258.66˚/s^2, *p* = 0.001) and significantly increased peak negative transverse plane trunk acceleration (Δ = 118.75˚/s^2, *p* = 0.003) were observed in the affected limb (Table 6.3). No other significant differences were observed (Appendix F, Table F.3, Table F.6, Table F.9).

Table 6.3 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the affected limb of the CAI group to the unaffected limb.

ANGULAR DISPLACEMENT	- No significant differences observed	- ↑ Forefoot-Tibia Inversion (affected: 7.32 ± 3.77, unaffected: 4.71 ± 4.74, p = 0.041)	- ↑ Forefoot-Tibia Inversion (affected: 7.33 ± 3.76, unaffected: 4.88 ± 4.49, p = 0.043) - ↓ Forefoot-Tibia Eversion (affected: 0.04 ± 3.89, unaffected: -2.21 ± 4.18, p = 0.018)
ANGULAR VELOCITY	- No significant differences observed	- No significant differences observed	- ↓ Forefoot-Tibia Inversion (affected: 4.00 ± 6.02, unaffected: 10.57 ± 11.48, p = 0.031) - ↑ Trunk External Rotation (affected: -2.68 ± 5.04, unaffected: 0.35 ± 5.23, p = 0.015)
ANGULAR ACCELERATION	- No significant differences observed	- No significant differences observed	- ↓ Forefoot-Tibia transverse peak +ve (affected: 1904.41 ± 697.36, unaffected: 2207.67 ± 848.05, p = 0.014) - ↑ Trunk sagittal peak –ve (affected: -475.74 ± 194.09, unaffected: -425.44 ± 192.41, p = 0.042) - ↓ Trunk frontal peak –ve (affected: -554.51 ± 224.99, unaffected: -813.17 ± 242.27, p = 0.001) - ↑ Trunk transverse peak –ve (affected: -479.51 ± 149.42, unaffected: -360.76 ± 171.36, p = 0.003)



6.3.4 Discussion

This study aimed to compare discrete kinematic variables that may be linked to injury likelihood between CAI participants' unaffected and affected ankles and to a matched control group. As significant differences were observed null hypothesis $H0_3$ can be rejected. The results of this study are comparable to a readily available published data set (Fukuchi et al., 2018).

CAI Group Affected Limb - Matched Control

Internal rotation is increasingly linked to increased strain and potential damage to the ATFL (Fong et al., 2009a; Fong et al., 2012; Konradsen & Voigt, 2002). Decreased FFTBA internal rotation of the CAI group's affected limb prior to HS may be a protective response in order to prevent strain on the ligament (Chinn et al., 2013), though this may also be due to joint restrictions (Vicenzino et al., 2006).

Decreased trunk lateral flexion towards the affected limb was observed as in previous research (Abdelraouf et al., 2012). Although not measured within this study, this may be an attempt to alter the position of the body's centre of mass. Significantly decreased FFTBA dorsiflexion velocity pre-HS, where the matched control group were dorsiflexing whilst the affected limb group demonstrated a negative plantarflexion velocity may place the ankle at increased susceptibility to LAS with plantarflexion being associated with ATFL sprain (Konradsen & Voigt, 2002).

Together, the significantly decreased FFTBA eversion angular acceleration, decreased FFHFA abduction angular velocity, and decreased FFTBA external rotation angular velocity observed in the affected limb at HS are of clinical importance as they identify that the velocity at which the affected limb of the CAI group moves out of the 'vulnerable position' of internal rotation and inversion is significantly decreased. This could be due to differences in peroneal reaction as found in previous studies (Delahunt et al., 2006a). Though it is unclear whether this is due to sensing the vulnerable position later followed by a normal duration reaction, or if this is due to the reaction itself being slower. Alternatively, this could be due to decreased proprioceptive control (Konradsen, 2002). Increased knee adduction angular velocity observed in the affected limb may suggest a proximal adaptation due to distal instability. Chinn et al. (2013) suggests that individuals with CAI may 'freeze' range at the ankle to maintain joint stability and therefore release degrees of freedom at the hip and/or knee joint. Similarly, Gribble et al. (2004) suggested that although the gross motor task may often be completed, the method of completion may be altered or less than optimal. The increased angular velocity of the knee in combination with the decreased angular velocities of the foot may together highlight the increased risk of recurrent LAS and episodes of giving way in the CAI affected limb. It is interesting to note that no significant differences occurred in angular displacement of these variables. Significantly decreased FFTBA eversion angular acceleration was observed in the affected limb at HS. Monaghan et al. (2006) observed an inversion velocity in CAI individuals during the period 5 ms pre-HS to 5 ms post-HS of 29 °/s whilst the healthy control group displayed an eversion velocity of 6 °/s. The CAI group in the present study showed decreased eversion velocity post-HS suggesting this may be an important area for the development of preventative measures.

As with the pre-contact phase, a decreased FFTBA internal rotation displacement and a decreased trunk lateral flexion angular displacement towards the affected limb was demonstrated in the affected limb. Increased trunk peak negative transverse plane acceleration and a decreased peak negative frontal plane acceleration were also observed, however, there is a paucity in research reporting these variables within individuals with CAI. Further research should investigate this finding as it is unclear what this shows.

CAI Group Unaffected Limb - Matched Control

Decreased trunk lateral flexion away from the stepping limb (towards the affected limb) observed pre-HS may be a protective mechanism to move the centre of mass away from the affected limb. Decreased FFTBA external rotation velocity was also observed which may be a result of increased rigidity of the FFTBA as a method of protecting the contralateral limb. Increased rigid and inflexible movement patterns have previously been found in the affected limb (Herb et al., 2014; Terada et al., 2015) however as far as the author is aware this is the first study to document this in the unaffected limb of individuals with CAI.

Significantly decreased FFHFA inversion and decreased FFTBA internal rotation angular displacement at HS may be an attempt to protect the affected limb by adopting a position of increased rigidity (Chinn et al., 2013). Interestingly an increased knee adduction angular velocity was also observed in the unaffected limb when compared to the matched control limb at HS, as was also the case between the affected limb and the matched control. Furthermore, no significant differences were observed in this variable between the CAI groups unaffected and affected limbs. This may highlight a possible cause for the high prevalence of bilateral ankle instability (Tanen et al., 2014).

Post-HS, a decreased trunk lateral flexion away from the stepping limb (towards the affected limb) was found, as in the period prior to HS, again suggesting a protective response to move the trunk away from the injured limb. There appears to be a paucity of research investigating trunk kinematics in lower extremity injuries, hence this finding warrants further investigation. Significantly decreased FFHFA eversion velocity was also observed. This implies that increased rigidity may be a protective mechanism for the contralateral limb but also suggests that the unaffected limb may be at increased risk of LAS as the period following HS is not within conscious control (Delahunt et al., 2006a; Monaghan et al., 2006). This suggests a decreased detection of inversion or a delayed peroneal muscle motor response as concluded previously (Monaghan et al., 2006).

CAI Group Affected Limb - CAI Group Unaffected Limb

At HS a significantly increased FFTBA inversion displacement was observed, placing the ankle in an increased position of vulnerability to LAS (Konradsen & Voigt, 2002). These findings have been previously reported when comparing the affected limb to a matched control (Delahunt et al., 2006a; Wright et al., 2013b), although this study does not report any significant differences for the same comparisons, probably due to differences in methodologies, and as such further research is warranted.

Post-HS increased FFTBA inversion and decreased FFTBA eversion angular displacement was observed in the affected limb suggesting a position of increased vulnerability to LAS, comparable to previous findings (Delahunt et al., 2006a; Drewes et al., 2009a; Monaghan et al., 2006). Decreased FFTBA peak positive transverse plane acceleration was observed in the affected limb. It is unclear what the implications of this are and this warrants further investigation. Decreased peak FFTBA inversion velocity in the affected limb may be a protective response as an attempt to increase rigidity within the joint (Herb et al., 2014; Terada et al., 2015).

A number of differences were observed in the trunk (increased trunk external rotation velocity, increased trunk peak negative acceleration (sagittal and transverse planes) and a decreased peak negative frontal plane acceleration, these combined suggest a significant modification in trunk kinematics though it is unclear whether this is caused by a deliberate modification or lack of control. The trunk accounts for approximately 35.5% of body mass therefore these changes are likely to be linked to changes in movement of the hip, knee and ankle (Kulas et al., 2008). Further research with external kinetic measurements from force plates and the use of an inverse dynamics model may help identify the causative factors for these differences along the kinetic chain.

Clinical Implications

The observation of reduced FFTBA transverse plane motion prior to initial contact in the affected limb when compared to the matched control is of particular clinical interest due to the close relationship of this variable to ankle sprains (Fong et al., 2009a; Fong et al., 2012; Konradsen & Voigt, 2002). The observed difference of 4.95˚ in internal rotation and 4.77˚ in external rotation may prove a clinically significant finding given a cadaveric study observed a maximal internal rotation displacement of 17.62˚ (Wilkerson, Doty, Gurchiek, & Hollis, 2010). Although it is unclear whether this reduced motion is due to a lack of range as observed by (Vicenzino et al., 2006) or whether this is a deliberate modification. This is a variable which may be a key area to focus future research. Equally the decreased FFTBA dorsiflexion angular velocity at the same time point may showing a change of 12.3˚/s may highlight an increased risk prior to initial contact.

Previous literature focuses on angular displacements however this research suggests that further significant differences may be prevalent in angular velocities and angular accelerations, and the variability of these may be the cause of, or a contributing factor to, instability. When planning rehabilitation programmes and injury prevention strategies, it is important to focus not just on the range but also the rate at which range is covered. Many rehabilitation exercises implemented in the late stage are slow in nature, however, these findings may call for more dynamic movements to be incorporated.

A number of studies have documented differences between the affected and the unaffected limb in CAI sufferers (De Ridder et al., 2013; Monaghan et al., 2006; Wright et al., 2013b), however, in this study a number of key significant differences between the unaffected limb and a matched control were observed suggesting rehabilitation should target both limbs.

Future Research

A prospective study is needed to determine whether the differences observed in the current study were present prior to or because of the initial injury. Future research should document joint angular velocities and angular accelerations with angular displacements as these may be more pertinent to episodes of giving way and mechanisms of injury. This method should also be used in more dynamic manoeuvres such as cutting and single-leg landing. It has been suggested that differences in angular displacement associated with CAI may be exacerbated during more dynamic movements (Donovan & Feger, 2017) and it would be interesting to investigate whether this is the case with angular velocities and accelerations. Lastly, future research should report kinetic parameters using this methodology in order to draw further conclusions from these preliminary findings.

6.3.5 Conclusion

Affected limbs of CAI sufferers appear to adopt a protective position with decreased internal rotation prior to and following HS. Modified transverse plane motion prior to initial contact may provide a key area for future research. Along with this a large number of modifications in angular velocity were also observed with a decreased FFTBA dorsiflexion velocity prior to HS and differences in FFTBA, FFHFA and knee velocity and acceleration present at HS which may increase the ankles susceptibility to recurrent LAS.

Altered movement patterns in the CAI group's unaffected limb may help to explain high bilateral CAI incidence rates.

6.3.6 Development of Research Within the Thesis

The present study highlighted several significant differences in angular displacement, angular velocity and angular acceleration. The next section of this thesis will use regression analysis to examine whether the significant kinematic variables observed during walking are able to predict scores on the IdFAI questionnaire

This knowledge will help in the development of preventative measures for ankle sprains but could also help to validate questionnaire use or alternatively developed as an objective marker for determining ankle instability.

6.4 Study 1 – Part 3 – Can Significantly Different Kinematic Variables Observed During Walking Between Individuals with and Without Ankle Instability Predict the Identification of Functional Ankle Instability Questionnaire Score?

6.4.1 Introduction

The previous study highlighted several differences that existed when comparing angular displacements, angular velocities and angular accelerations prior to HS, at HS and following HS.

No globally accepted measure has been agreed upon for the diagnosis of ankle instability (Simon et al., 2014) and subsequently, the use of questionnaires in the reporting and classification of ankle instability has been disputed due to the reliance on a self-reporting nature. Reliability has been performed on each questionnaire to identify appropriate cut off values (Gurav et al., 2014; Simon et al., 2014) and as such, these are reported in the international ankle consortium (IAC) guidelines for CAI selection criteria (Gribble et al., 2013). The IdFAI questionnaire provides a score between 0 and 37, however, the IAC guidelines suggest a cutoff score of 11 or more indicates instability. This means someone who scores a 10 would be determined 'healthy' whilst someone with a score of 11 is deemed to have ankle instability. This study will look to address whether differences observed between groups in kinematic walking variables could be used to predict the score of the IdFAI questionnaire.

This study aims to identify whether it is possible to predict scores on the IdFAI questionnaire from the significantly different variables observed during walking.

This study will address the following hypothesis previously presented in Chapter 2.19:

- **H_4** - Significant differences observed during walking will be able to predict IdFAI questionnaire score

6.4.2 Methods

Variables where significant differences were observed between the affected limb of the CAI group and the healthy matched control limb during walking (study 1, part 2 (6.2.4)), were used for analysis. These are outlined in Table 6.4.

Table 6.4 Variables with significant differences in study 1 part 2 (6.3.3) used within the regression analysis.

	100 ms pre- heel strike	**Heel strike**	**Heel strike – 200 ms post heel strike**
Angular displacement	- ↓ FFTBA internal rotation - ↓ FFTBA external rotation - ↓ Trunk lateral flexion (towards the affected limb)		- ↓ FFTBA internal rotation - ↓ Trunk lateral flexion (towards the affected limb)
Angular velocity	- ↓ FFTBA dorsiflexion	- ↓ FFHFA Abduction - ↓ FFTBA external rotation - ↑ Knee Adduction	- ↓ FFTBA external rotation
Angular acceleration	- ↑ FFTBA transverse peak +ve - ↓ Knee frontal peal –ve	- ↓ FFTBA eversion	- ↓ Trunk frontal peak –ve - ↑ Trunk transverse peak -ve

Statistical Analysis

Normality was assessed using the Shapiro-Wilks test. Due to the sample size used, Pearson's correlation analysis was performed to identify whether relationships between

IdFAI questionnaire score and kinematic variables existed. All predictors that showed a moderate (r = 0.3 upwards; Cohen & Cohen, 1977) and significant correlation ($p < .05$) were kept for further analysis, while all other variables were removed from subsequent analysis. Linear or multiple stepwise regressions were then used to identify which kinematic variables during walking best predicted IdFAI score. The IdFAI score was the criterion and the kinematic variables the independent variables. Independent variables were examined for co-linearity prior to entry into the regression model and those with high co-linearity $R \geq 0.7$ were removed. Statistical analysis were performed in SPSS 24.0 (SPSS Inc, Chicago, USA). Data were inspected to ensure assumptions of linearity, independence of errors, homoscedasticity and normality of residuals to ensure it met all assumptions of a regression before proceeding.

6.4.3 Results

Significant correlations can be observed in Table 6.5. Of the 14 variables, 11 variables were inputted into a stepwise regression analysis.

Table 6.5 Pearson's correlation outputs for kinematic variables compared to IdFAI score

		FFTBA transverse internal rotation displacement 100 HS	FFTBA transverse ER displacement 100 HS	Trunk frontal lateral flexion displacement 100 HS	FFTBA transverse +ve acceleration 100 HS	FFTBA sagittal DF velocity 100 HS	FFTBA frontal acceleration HSVALUE	FFHFA transverse abduction velocity HSVALUE	FFTBA transverse ER velocity HSVALUE	Knee frontal adduction velocity HSVALUE	FFTBA transverse internal rotation displacement HS 200	Trunk frontal lateral flexion displacement HS 200	Trunk frontal +ve acceleration HS 200	Trunk transverse -ve acceleration HS 200	FFTBA transverse ER velocity HS 200
IdFAI Score	Pearson Correlation	-0.482*	-0.463*	0.428*	0.342*	-0.1687	0.344*	0.280	0.385*	0.402*	-0.416*	0.346*	0.462*	-0.321	0.383*
	Sig. (2-tailed)	0.003	0.004	0.009	0.041	0.325	0.040	0.099	0.021	0.015	0.012	0.039	0.005	0.056	0.021

The stepwise regression analysis showed FFTBA transverse plane internal rotation displacement prior to HS was the best independent predictor of IdFAI score (R = 0.482, R^2 = 0.232, F = 10.285, p = 0.003). Combined FFTBA transverse plane internal rotation displacement prior to HS and FFTBA transverse +ve acceleration prior to HS improved prediction (R = 0.648, R^2 = 0.420, F = 10.683, p = 0.003). The final model produced included trunk lateral flexion (towards the affected limb) displacement following HS (R = 0.705, R^2 = 0.496, F = 4.858, p = 0.0035). Prediction equations are outlined in Table 6.6 and Beta and standard error values are outlined in Table 6.7.

Table 6.6 Prediction equations for IdFAI score

Model	Equation
Model 1	IdFAI Score = (-0.925 x FFTBA transverse internal rotation displacement 100 HS) + 16.221 (± 8.479)
Model 2	IdFAI Score = (-1.074 x FFTBA transverse internal rotation displacement 100 HS) + (0.542 x FFTBA transverse +ve acceleration 100 HS) + 9.805 (± 7.480)
Model 3	IdFAI Score = (-1.015 x FFTBA transverse internal rotation displacement 100 HS) + (0.416 x FFTBA transverse +ve acceleration 100 HS) + (1.457 x Trunk lateral flexion displacement HS 200) + 12.318 (± 7.078)

Table 6.7 Unstandardized and standardized Beta values for each of the 9 regression models

Dependent variable	Variable	B	SE	ß	st error of estimate
IdFAI Score (model 1)	Constant	16.221	1.973		
	FFTBA transverse internal rotation displacement 100 HS	-0.925	0.288	-0.482	8.479
IdFAI Score (model 2)	Constant	9.805	2.624		
	FFTBA transverse internal rotation displacement 100 HS	-1.074	0.259	-0.559	
	FFTBA transverse +ve acceleration 100 HS	0.542	0.166	0.440	7.480
IdFAI Score (model 3)	Constant	12.318	2.732		
	FFTBA transverse internal rotation displacement 100 HS	-1.015	0.246	-0.528	
	FFTBA transverse +ve acceleration 100 HS	0.416	0.167	0.338	
	Trunk frontal +ve displacement HS 200	1.457	0.661	0.294	7.078

6.4.4 Discussion

The IdFAI questionnaire is often used by clinicians and within research for the identification of ankle instability. The questionnaire places individuals onto a scale. From this scale, anyone with a score of 11 or higher is classed to have ankle instability and somebody with a score of 10 would be deemed healthy. This study follows on from the results of study 1 part 2 where variables that were deemed to be significantly different between groups were used to identify whether they can determine the 'degree' of instability on the IdFAI questionnaire.

Of the 14 variables included within this analysis, 11 were significantly correlated to the IdFAI questionnaire score. Questionnaires have been questioned in the determination of ankle instability (Donahue et al., 2011), potentially, due to the self-reporting nature and thus the impact of the biopsychosocial model on interpretation of disability and the impact of pain-related fear or kinesiophobia (Lentz, Sutton, Greenberg, & Bishop, 2010). Due to this criticism, the use of additional variables may help to strengthen its use. The regression analysis produced 3 models. The first used FFTBA transverse plane internal rotation displacement prior to HS and was able to predict 23.2% of the variance. The higher score on the IdFAI questionnaire seemed to result in less internal rotation. Internal rotation is a known mechanism for ankle sprains (Fong et al., 2009a; Fong et al., 2012; Konradsen & Voigt, 2002). The reduced internal rotation observed prior to HS may be a preventative measure in an attempt to increase stability as a type of feedforward strategy. However, this may also predispose individuals to further injuries by placing the ankle in a less 'closed-packed' and more of a loose-packed position.

The second model included FFTBA transverse plane positive acceleration prior to HS. This increased the IdFAI score prediction percentage to 42%. The correlation showed as the IdFAI score increases so too does the positive acceleration. It again seems logical that the movement occurs at the foot and prior to HS suggesting the position adopted pre-IC may be the best predictor of ankle instability.

The final model also incorporated trunk lateral flexion displacement (towards the affected limb) from HS to 200 ms post HS increasing the percentage to 49.6%. The Pearson's correlation seemed to show the more unstable the limb the more they moved away from the injured limb. It is possible that this is a protective mechanism in order to change the position of the centre of gravity away from the injured leg towards the contralateral limb. As models were produced that were able to predict the IdFAI questionnaire score null hypothesis $H0_4$ can be rejected.

Clinical Implications

Further research should be conducted to confirm whether the variables listed above may also be used to increase the specificity and sensitivity of tests to confirm ankle instability. These variables, however, may be key when implementing rehabilitation and preventative strategies. Again, further research is needed to determine whether these predisposed individuals to the initial ankle injury or whether these are subsequent adaptations that have occurred as a result of the initial injury. It is also necessary to conduct the same type of research with other key sporting movements to determine whether these variables are similar across movements.

6.4.5 Conclusion

Forefoot-tibia transverse internal rotation displacement and FFTBA transverse peak positive acceleration prior to HS along with trunk lateral flexion displacement post HS were able to account for 49.2% of the IdFAI score variance. This shows a high prediction ability with just three variables, but further research is needed in other movements.

6.4.6 Development of Research Within the Thesis

Following on from the work of this chapter, the next chapter will aim to gain further insight into movement patterns and muscle activation patterns during single-leg landing; a more dynamic mechanism often observed as a mechanism for ankle sprains. Obtaining more information during a range of movements is crucial, to identify key areas that should be addressed in rehabilitation and injury prevention and to further understand possible causes or predisposing factors for recurrent ankle sprains in those with ankle instability.

Chapter 7.0 Study 2 – Analysis of Single-Leg Landing Muscle Activation and Kinematics in Individuals with Ankle Instability and Healthy Control Participants

7.1 Single-Leg Landing Chapter Overview

Ankle sprains are a frequent occurrence in sports involving rapid changes of direction, jumping and landing (Fong et al., 2008). Of the 39% of noncontact ankle sprains incurred in English Premier League football (July 1997 - end of May 1999), landing was described as the mechanism of injury for 36% of noncontact ankle sprains (Woods et al., 2003). Individuals who suffer recurrent ankle sprains, episodes of giving way and feelings of instability (chronic ankle instability) have been shown to have both decreased activity levels (Hubbard-Turner & Turner, 2015) and an increased risk of early-onset post-traumatic osteoarthritis of the ankle joint (Valderrabano et al., 2006). As such, enhanced understanding of differences in kinematic movement patterns of those with CAI when performing landing manoeuvres may be beneficial to prevention and rehabilitation strategies.

Landing can be broken down into the pre-landing, impact and the reactive phases of landing (Doherty et al., 2014; Lees, 1981). The period before IC can be termed pre-landing. This phase of landing is thought to be feed-forward motor control, where individuals use pre-programmed mechanisms recruited in order to modify their centre of gravity in an attempt to maintain ankle joint stability and anticipate the imminent joint loading with ground contact (Delahunt et al., 2007). Muscle activity pre-empting landing is suggested to occur at approximately 200 ms pre-ground contact (Santello, 2005). This response is thought to be modulated by vision (Santello, 2005), and therefore may play a particular role in injury prevention. Landing from a jump takes place over approximately 1 second, however, the impact absorption phase is known to last for 150-200 ms (Lees, 1981). Beyond this point, downwards momentum is reduced and the rest of the action is concerned with the maintenance of balance (Lees, 1981). This phase is referred to as the reactive phase of landing (Doherty et al., 2014). It has been summarised that an ankle sprain can occur as early as 40 ms after IC (Fong et al., 2009b). This phase is beyond human control and instead is based on system reaction and stabilisation rather than a conscious modification of movement (Lees, 1981).

Altered foot positioning has been suggested to predispose individuals with CAI to recurrent ankle sprains. A study reporting kinematics of an accidental ankle sprain occurring during a laboratory-based landing manoeuvre observed greater ankle inversion and internal rotation angles during the pre-landing phase and at IC (Y. Li et al., 2018). They also suggested that altered hip mechanics may have resulted in unanticipated timing for ground contact or modified sensation in the knee and ankle joints resulting in giving way (Y. Li et al., 2018). However, it is unclear whether individuals with CAI would consistently display modified movement mechanics compared to healthy individuals.

Previous literature investigating CAI during landing has used single segment foot models (De Ridder et al., 2015a; Delahunt et al., 2006b, 2007), however, this may lead to unrepresentative conclusions being drawn in relation to the biomechanics of the foot (Stebbins et al., 2006). De Ridder et al. (2015b) appears to be the first study to analyse single-leg landing using a multi-segmental foot model. They compared the use of the Ghent Foot Model to a rigid foot model in participants with CAI, copers (defined in the study as those who have suffered from a recent ankle sprain but no symptoms of instability) and control participants. The single segment foot model showed a less inverted position from 10-100% of the impact phase in the CAI group and the coper group when compared to the control group. Comparatively the only differences observed in the multi-segmental model were in the hallux segment. It was suggested that the single segment foot model over simplified findings not documenting the movement of the hindfoot, and as such results should be interpreted with caution (De Ridder et al., 2015b).

The trunk segment mass and the distribution of this mass is known to contribute greatly to ground reaction forces during landing (Kulas et al., 2008). As the kinetic chain principle suggests, it is thought that the movement of this segment will influence movements of the lower extremity. Further to this, potential relationships have been proposed with proximal adaptations to injury (Doherty et al., 2016a).

Following the methodology adopted in Chapter 6.0, this chapter will adopt a similar 3 part structure to further analyse differences in muscle activation and movement kinematics. Chapter 7.2 will analyse full time series movement patterns and muscle activation patterns between individuals with CAI and healthy controls during a single leg landing. Chapter 7.3 will address the angular displacements, angular velocities and angular accelerations displayed by groups during a single-leg land. Lastly, Chapter 7.4 will examine whether the findings observed in Chapter 7.2 can be used as a prediction of the individuals IdFAI questionnaire score.

7.2 Study 2 – Part 1 – Analysis of Lower Limb and Trunk Kinematics and Muscle Activations During Single-Leg Landing in Participants with and Without Ankle Instability

7.2.1 Introduction

The majority of research analysing single-leg landings in CAI populations reports joint angles and muscle activation characteristics at discrete time points (Caulfield & Garrett, 2002; Delahunt et al., 2006b). However they do not report kinematic parameters for the whole time-series curve, potentially missing significant differences or trends (Pataky et al., 2013). Statistical parametric mapping enables curve analysis across the whole movement (Pataky et al., 2013). De Ridder et al. (2015a; 2015b) used statistical parametric mapping to compare single-leg landing foot kinematics between participants with CAI, copers, and controls, identifying exact time periods of significance within the impact phase of landing. This study will combine trunk kinematic analysis with a full lower limb and multi-segmental foot model and analysis of proximal and distal muscle activation patterns to compare single-leg landing strategies of individuals with and without CAI.

The aim of this study was to compare trunk, hip, knee, and multi-segmental foot kinematics and muscle activation during single-leg landing; for the pre-landing and impact phases of movement between participants with CAI and healthy controls to highlight potential differences that may be addressed for the development of intervention and prevention strategies.

This study will address the following hypotheses previously presented in Chapter 2.19:

- $\mathbf{H_5}$ - CAI participants will display modified kinematic movement patterns (SPM) during single-leg landing
- $\mathbf{H_6}$ - CAI participants will display modified muscle activation patterns (SPM) during single-leg landing

7.2.2 Methods

Eighteen healthy controls (14 males, 4 females, age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 participants with CAI (13 males, 5 females, age: $\bar{\chi}$ = 22.0 ± 2.7 years, height: $\bar{\chi}$ = 176.8 ± 7.9 cm, mass: $\bar{\chi}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were included in this study as outlined in General Methods 3.1.

Protocol

Participants performed a 5-minute warm-up on a cycle ergometer (Monark Ergomedic 874E, Sweden) at 60 Watts. Electromyographic data were recorded for the gluteus medius, peroneus longus and tibialis anterior as outlined in general methods Chapter 3.4. Electromyographic data were normalised to the activity mean for each landing (Bolgla & Uhl, 2007). Motion analysis data were recorded as outlined in general methods Chapter 3.5.

Participants were required to perform three barefoot single leg drop landings onto each limb from a 30 cm high box (Kunugi et al., 2017) onto a flat stable laboratory floor with 1-minute rest between trials. The order of trials was randomised to minimise the effect of fatigue. Individuals were asked to hop forward off the box onto the floor in front and maintain balance for 3 seconds whilst looking straight forward. No instruction was given to participants regarding arm position during the landing manoeuvre in order to observe an unmodified landing position. Trials were discarded if the second foot contacted the floor to restore balance or if errors were observed in marker tracking. Each participant performed a familiarisation of the movement until they were comfortable with the movement before recording.

Data and Statistical Analysis

Data were inspected using Cortex (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) software before importing into Visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland). Data were smoothed using a 6 Hz Butterworth filter. Initial contact was determined using peak plantarflexion forefoot-tibia velocity (following this point dorsiflexion occurred). Event markers were created for IC, 200 ms pre-IC and 200 ms post- IC. EMG data were root mean squared by a moving window of 100 ms and normalised to mean task activation. Following this, a visual inspection of the data identified noise in the signal for two of the participants that warranted their EMG data be removed. To keep the pre-experimental research design, the matched controls assigned to the two participants also had their EMG data removed. Kinematic and EMG data were exported for the pre-landing (200 ms pre-IC to IC) and impact (IC to 200 ms post-IC) phases into MATLAB R2015a (The Math Works, Natick, Massachusetts) to perform the SPM analysis.

Kinematic data were exported for the forefoot-hindfoot angle, forefoot-tibia angle, hindfoot-tibia angle, hip, knee and trunk angles in the sagittal, frontal and transverse planes of motion. So not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment (De Ridder et al., 2013; Wright et al., 2011). Data were analysed using SPM (Friston et al., 1994) in MATLAB using the SPM1D open-source package (spm1d.org).

Normality tests were performed using a D'Agostino-Pearson's test. A matched control limb was compared to the CAI groups affected limb and unaffected limbs using an independent samples *t*-test ($\alpha = 0.05$). The unaffected and affected limb of the CAI group were then compared using a paired samples *t*-test ($\alpha = 0.05$).

7.2.3 Results

CAI Group Affected Limb Versus Matched Control

Independent samples *t*-tests showed no significant differences in FFHFA (Appendix H, Figure H.1, Figure H.8), FFTBA (Appendix H, Figure H.2, Figure H.9), HFTBA (Appendix H, Figure H.3, Figure H.10), hip (Appendix H, Figure H.4, Figure H.11), knee (Appendix H, Figure H.5, Figure H.12) or trunk (Appendix H, Figure H.6, Figure H.13) angles in the frontal, sagittal or transverse planes or gluteus medius, peroneus longus or tibialis anterior muscle activation patterns during the pre-landing or impact phases (Appendix H, Figure H.7, Figure H.14).

CAI Group Unaffected Limb Versus Matched Control

Independent samples *t*-tests showed significantly increased hip abduction in the unaffected limb when compared to the matched control at 30-100% of the impact phase of landing ($p = 0.011$, mean difference = 4.28°, peak difference = 4.47°) (Figure 7.1) though no significant differences were observed in this variable during the pre-landing phase (Appendix H, Figure H.32). No significant differences were observed in FFHFA (Appendix H, Figure H.29, Figure H.36), FFTBA (Appendix H, Figure H.30, Figure H.37), HFTBA (Appendix H, Figure H.31, Figure H.38), knee (Appendix H, Figure H.33, Figure H.39) or trunk (Appendix H, Figure H.34, Figure H.40) in the frontal, sagittal or transverse planes in gluteus medius, peroneus longus or tibialis anterior muscle activation patterns (Appendix H, Figure H.35, Figure H.41) during the pre-landing or impact phases.

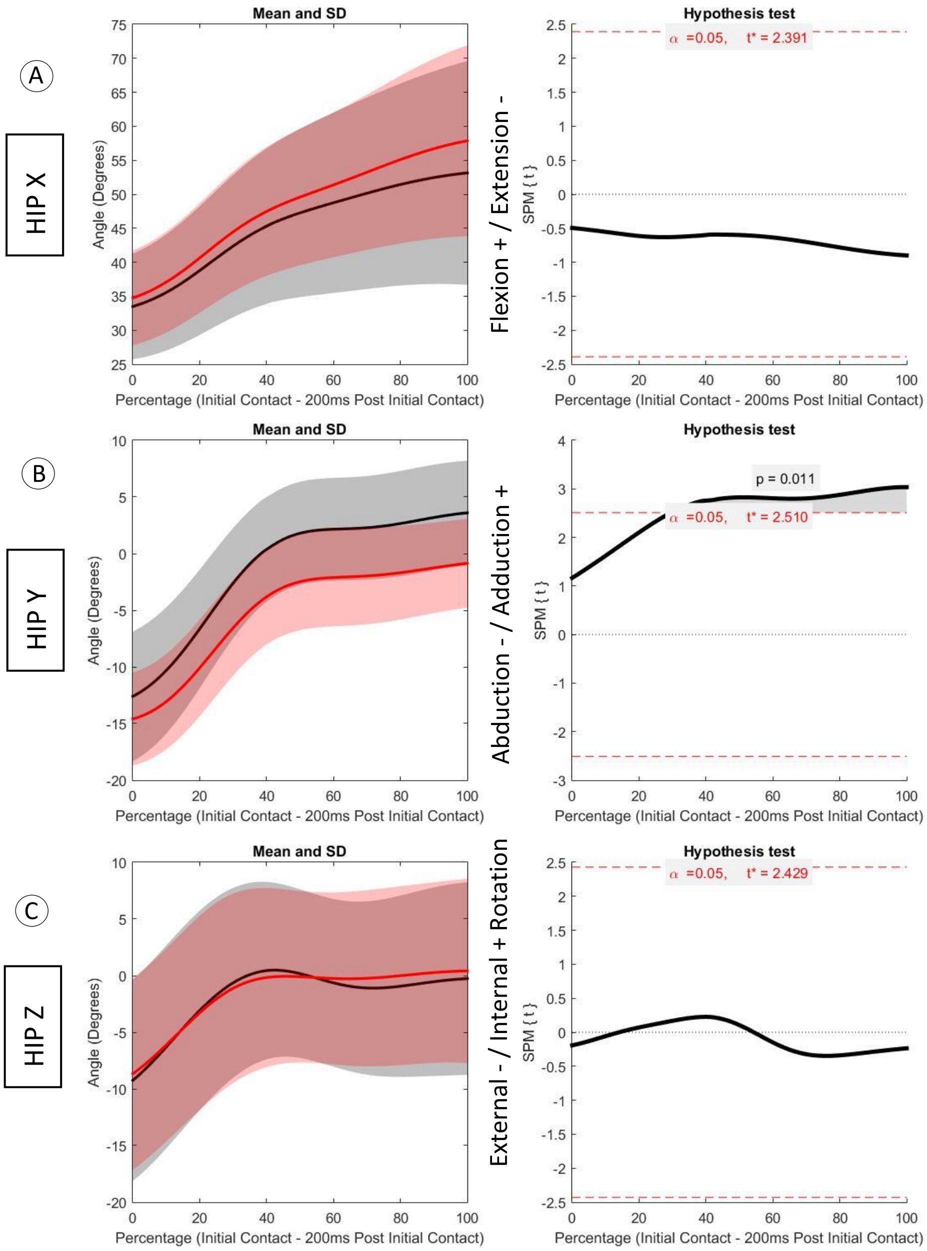


Figure 7.1 Hip angles (x, y, z) IC to 200 ms post-IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

CAI Group Affected Limb Versus CAI Group Unaffected Limb

Paired samples *t*-tests showed no significant differences in FFHFA, FFTBA, HFTBA, hip, knee or trunk in the frontal, sagittal or transverse planes or in gluteus medius, peroneus longus or tibialis anterior muscle activation patterns from 200 ms pre-IC - IC or from IC to 200 ms post-IC (Appendix H, Figure H.15-Figure H.28).

7.2.4 Discussion

This study aimed to provide a comprehensive analysis of the trunk and lower limb during landing utilising a multi-segmental foot model. It also aimed to compare muscle activation patterns between athletes with CAI and healthy controls. It is the first of its kind to not only compare the affected limb to a matched control, but also to compare between the affected and unaffected limbs of the CAI group and to compare between the unaffected limb of the CAI group and a matched control for kinematic analysis of the trunk and lower limb combined with a multi-segmental foot model along with lower limb muscle activation patterns. As significant differences were observed in kinematics null hypothesis $H0_5$ can be rejected. As no significant differences were observed in muscle activation $H0_6$ cannot be rejected.

Significantly increased hip abduction was observed in the unaffected limb when compared to the matched control at 30-100% of the impact phase of landing. Although this difference is small (mean difference 4.28°) it is interesting to note that the CAI group appear to abduct the hip moving away from the affected limb potentially to avoid touchdown of the affected limb. Previous literature investigating gender differences in landing suggested an increase in hip abduction during touch down allows for the hip to move through a greater range of motion towards adduction and also keeps the gluteus medius closer to its resting length, therefore, allowing an increased control of deceleration (Weinhandl, Joshi, & O'Connor, 2010). This is suggestive of a hip strategy and evidence of proximal adaptations of the kinetic chain within the contralateral limb. With an additional external stimulus alongside fatigue or increased movement velocity, this may be a potential risk factor. This is the first study to document differences in kinematic movement patterns of the unaffected limb.

No significant differences were observed between affected limb and the matched control group for hip kinematics in this study. This is in line with a previous study analysing the full movement trace (De Ridder et al., 2015a) and others reporting discrete variable data (Delahunt et al., 2007; Gribble & Robinson, 2009, 2010). However, one study observed a less externally rotated hip position from 200 ms – 55 ms pre-IC (Delahunt et al., 2006b). The authors suggested this may provide evidence that neuromuscular impairments may not be confined to just the ankle joint and that potential central neural adaptions may exist with peripheral joint issues. The present study suggests there are no differences in the pattern of movement adopted in the affected limb of the CAI group when compared to a matched control group during the landing and pre-landing phase.

Literature in landing mechanics has previously documented the involvement of the trunk in landing kinematics (Lees, 1981). The author felt it was pertinent to explore the kinematics of this segment, however, no significant differences were observed in trunk kinematics between groups.

No significant differences were observed in knee kinematics in the sagittal, frontal or transverse planes. These findings are in line with a study also using full curve SPM analysis (De Ridder et al., 2015a) and they are consistent with studies that have not performed SPM analysis (Delahunt et al., 2006b, 2007). Caulfield et al. (2002). found increased knee flexion in the CAI group from 20 ms pre-IC to 60 ms post-IC when performing a drop jump from a height of 40 cm. This was attributed to central patterning at the spinal level during landing. In contrast, research analysing a maximal double leg vertical jump with a single-leg landing observed increased knee flexion prior to IC in the control group which was proposed to assist with stability by lowering the centre of gravity (Gribble & Robinson, 2009, 2010). These differing findings may be due to slight differences in the movements being analysed, however they make drawing substantial conclusions for clinical practice and intervention strategies difficult.

Foot and ankle kinematics are often reported in the literature with the use of a single segment foot model, however, research comparing results for single-leg landing comparing a multi-segmental and a single segment foot model reported that results of the single segment model should be interpreted with caution and that the use of a multi-segmental model may highlight further differences between groups in distal kinematics (De Ridder et al., 2015b). In this study, no significant differences were observed in foot and ankle kinematics. This is consistent with findings from De Ridder et al. (2015b) who reported no significant differences in hindfoot and midfoot values. This leaves the reason for the increased susceptibility of ankle sprain during landing in individuals with CAI unclear. De Ridder et al. (2015b) observed an increased peak vertical ground reaction force, decreased time to peak and an increased loading rate when compared to the control group. These parameters were not recorded in the present study. Further research should be conducted investigating other kinematic factors such as joint angular acceleration and velocity to observe the role that these play in the landing mechanism within individuals with CAI.

No significant differences were observed in muscle activation patterns of the gluteus medius during the single limb landing. This concurs with previous findings investigating lateral hops pre and post-fatigue that found no significant differences between groups for muscle activation (K. A. Webster et al., 2016). Similarly, no significant differences were observed in tibialis anterior muscle activation patterns in the present study. When investigating double leg landing with the test limb landing on a tilted force plate, increased tibialis anterior muscle activations were observed in participants with CAI (Y. Li et al., 2017). The authors suggested that this increased dorsiflexor activation is an attempt to stabilise the ankle joint (closed packed position) to limit ankle inversion displacement and eversion moment. This was particularly necessary on the inverted surface used within this study. Previously increased peroneal longus muscle activation has been observed with drop jump landing (Herb, Grossman, Feger, Donovan, & Hertel, 2018), however, this study also found differences in ankle kinematics. Within the present study, no differences were observed in foot kinematics, therefore, this may explain why no significant differences were observed in tibialis anterior and peroneus longus activation patterns in the present study.

This study analysed kinematic and EMG parameters to examine movement patterns and muscle activations patterns during single-leg landing between groups. The statistical analysis method implemented involves time normalisation and as such is referring to the pattern of movement and does not address possible differences that may be present in angular velocity and angular acceleration. It is also key to note that although symptoms such as giving way, instability and recurrent ankle sprains are attributes to CAI, these are not continuous and therefore differences in the recorded parameters may not be consistently present.

Future research should investigate angular velocities and angular accelerations to determine whether any significant differences are present. Investigation of the impact of CAI on kinetic parameters using full curve analysis to identify if differences exist in movement and muscle activation patterns between groups should also be performed. Kinetic analysis should further investigate differences between the unaffected limb and the matched control. This study observed increased hip abduction in the unaffected limb at 30-100% of the impact phase of landing. Reporting the position of the centre of mass would be beneficial as it is postulated that this may be an attempt to shift the centre of mass away from the affected limb. In doing so this may place the unaffected limb at increased risk of lateral ankle sprains. Reference to kinetic parameters may highlight differences between groups that were not evident in the current study.

Clinical Implications

No significant differences were observed in movement patterns adopted by the affected limb when compared to the matched control limb or when comparing the affected limb to the unaffected limb of the CAI group. When comparing the unaffected limb to the matched control limb however an increased hip abduction was observed at 30-100% of the impact phase of landing with a mean difference of 4.28 degrees and a peak difference of 4.47 degrees. Abnormal femoral motion has been reported to have a direct impact on tibiofemoral joint kinematics (Powers, 2010). In addition to this the relative locations of the centre of mass and the centre of pressure will impact the orientation of the resultant ground reaction force vector in relation to the ankle (Powers, 2010) – which may predispose the unaffected limb to lateral ankle sprain or other lower extremity injuries. This may also be evidence of a compensated Trendelenburg sign moving the resultant ground reaction force closer to the hip joint centre thus reducing the demand that is placed upon the hip abductors. This highlights the importance of bilateral intervention strategies and an area for future research to be targeted.

7.2.5 Conclusion

A significantly increased hip abduction was observed in the unaffected limb of the CAI group when compared to a matched control at 30-100% of the impact phase. This is potentially a protective method to move away from the affected limb. Further research should be conducted incorporating kinetic analysis. No other significant differences were observed in the trunk, hip, knee, FFTBA, FFHFA or HFTBA kinematics in the sagittal, frontal or transverse planes of motion or in muscle activation patterns between groups. Differences between groups that predispose individuals with CAI to recurrent ankle sprains and the feeling of instability and/or giving way, may not consistently manifest themselves in changes to joint angular displacements during single limb landing.

7.2.6 Development of Research Within the Thesis

Study 1 part 1 investigated similar analysis to this study during walking. In this study, a significantly increased FFTBA inversion was observed in the affected limb when compared to the unaffected at 4-16% of the stance phase. No significant differences were observed in inversion between groups. Interestingly, an increased hip abduction was observed during the impact phase of landing in the unaffected limb when compared to the matched control whereby the limb was abducted away from the affected limb. This is a particularly interesting finding and warrants further investigation in more dynamic tasks such as cutting. As with the previous chapter, there is also a call for investigating kinematic velocities and accelerations as these may also significantly impact risk factors for ankle sprains.

7.3 Study 2 – Part 2 - A Comparison of Lower Limb Angular Displacements, Angular Velocities, and Angular Accelerations During Single-Leg Landing Between Participants with Chronic Ankle Instability and Healthy Controls

7.3.1 Introduction

Current literature analysing landing kinematics has focused on angular displacements (Caulfield & Garrett, 2002; De Ridder et al., 2015a; De Ridder et al., 2015b). Research analysing individuals with CAI using a tilt platform has found the same degree of movement to be covered in a significantly shorter period of time (Vaes et al., 2001). This suggests that the rate at which a range is covered may be a more prominent predisposing factor associated with CAI individuals. An increased speed during loading has been found to correlate with increased ligament stress which leads to ligament sprain (Vaes et al., 2001). Decreased sagittal plane dorsiflexion velocity has been reported using a single segment foot model prior to IC showing that the time taken to move from the open-packed vulnerable position to a more stable closed-packed position was significantly increased have been reported prior to IC (Delahunt et al., 2006b). Accepted limits for ankle inversion velocities have been published for during running, cutting and landing manoeuvres. It is suggested that velocities should be less than 300 °/s (Chu et al., 2010). Injuries have been reported with velocities approximately 600 °/s and higher (Chu et al., 2010). These limits have, however, been established using a single segment foot model, so it is unclear whether these limits would transfer to a multi-segmental foot model. Minimal research currently exists investigating angular acceleration during landing. During trials analysing accidental ankle sprains sustained during laboratory-based cutting analysis increased plantarflexion and internal rotation angular velocity were observed in the injury trial (Fong et al., 2009a). Similarly, Kristianslund et al. (2011)

reported significantly increased inversion angular velocity at the ankle joint during an accidental ankle sprain during a cutting trial. Displacement data alone may not highlight all differences between participants with and without CAI, angular velocities and angular accelerations may provide further insight.

The aim of this study was to compare angular displacement, velocities and accelerations of the trunk and lower limb from the sagittal, frontal and transverse planes during the pre-landing, IC and impact phases of landing between participants with CAI and healthy controls.

This study will address the following hypothesis previously presented in Chapter 2.19:

- **H_7** - CAI participants will display modified discrete kinematic variables during single-leg landing

7.3.2 Method

Participants

Eighteen healthy controls (14 males, 4 females, age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 participants with CAI (13 males, 5 females, age: $\bar{\chi}$ = 22.0 ± 2.7 years, height: $\bar{\chi}$ = 176.8 ± 7.9 cm, mass: $\bar{\chi}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were included in this study as outlined in General Methods 3.1.

Protocol

Kinematic data from the single leg landing trials collected in Chapter 7.2 were inspected in Cortex (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) and imported into Visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland, USA) for analysis within this study.

Data Reduction and Statistical Analysis

Data were smoothed using a 6 Hz Butterworth filter. Initial contact was determined using peak plantarflexion forefoot-tibia velocity (following this point dorsiflexion occurred). Event markers were created for IC, 200 ms pre-IC and 200 ms post-IC. Kinematic data were exported for peak angular displacement, angular velocity and angular acceleration at 200 ms pre-IC to IC (pre-landing), IC and IC to 200 ms post-IC (impact phase). Kinematic data were exported as metrics for the forefoot-hindfoot angle, forefoot-tibia angle, hindfoot-tibia angle, hip, knee and trunk angles in all planes of motion. So not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment (De Ridder et al., 2013; Wright et al., 2011). Statistical analysis was performed in SPSS (IBM Corp. Released 2016. IBM SPSS Statistics for Windows, Version 24.0. Armonk, NY: IBM Corp). The matched control limb was compared to the CAI group's affected limb using an independent samples *t*-test. The unaffected and the affected limb of the CAI group were then compared using a paired samples *t*-test ($\alpha = 0.05$). A matched control limb was also compared to the CAI group's unaffected limb using an independent samples *t*-test ($\alpha = 0.05$).

7.3.3 Results

Independent samples *t*-tests for full kinematic curve analysis revealed no significant differences ($p > 0.05$) between groups for age, height, mass, or shoe size.

CAI Group Affected Limb Versus Matched Control

When investigating angular displacement during the pre-landing phase a significantly decreased FFTBA internal rotation angular displacement (Δ = 3.89˚, p = 0.046) was observed in the affected limb of the CAI group (Table 7.1). In angular velocity, a significantly increased HFTBA eversion angular velocity (Δ = 20.89˚/s, p = 0.048), a significantly increased knee adduction angular velocity (Δ = 18.25˚/s, p = 0.037) and an increased trunk flexion velocity (Δ = 21.38˚/s, p = 0.025) were observed in the affected limb of the CAI group when compared to the matched control (Table 7.1). Increased peak negative hip frontal plane angular acceleration (Δ = 617.50˚/s^2, p = 0.035) and increased peak negative trunk sagittal plane angular acceleration (Δ = -437.26˚/s^2, p = 0.038) was observed in the affected limb when compared to the healthy matched control in the pre-landing phase (Table 7.1). No other significant differences were observed between the affected limb and the matched control pre-landing (Appendix I, Table I.1, Table I.4, Table I.7).

At IC, no significant differences were observed in angular displacements between groups (Appendix I, Table I.2). A significantly increased HFTBA eversion velocity (Δ = 60.43˚/s, p = 0.004) and a significantly increased trunk flexion angular velocity (Δ = 26.10˚/s, p = 0.017) were observed in the affected CAI group (Table 7.1). A significantly increased negative knee sagittal plane angular acceleration was observed in the affected limb (Δ = 1363.22˚/s^2, p = 0.018) when compared to the matched control at IC (Table 7.1). No other significant differences were observed at this time point for angular velocity and angular acceleration (Appendix I, Table I.5, Table I.8).

During the post-impact phase, a significantly increased peak knee flexion velocity (Δ = 50.75˚/s, p = 0.007) and increased peak knee external rotation velocity (Δ = 2.77˚/s, p =

0.048) was observed in the affected limb. Increased peak negative knee sagittal plane angular acceleration (Δ = -1257.98°/s^2, p = 0.011) was also observed in the affected limb during the post-impact phase (Table 7.1). No other significant differences were observed during the post impact phase of landing between the affected and the matched control limb (Appendix I, Table I.3, Table I.6, Table I.9).

Table 7.1 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the affected limb of the CAI group to a matched control.

AFFECTED COMPARED TO MATCHED CONTROL			
	200 MS PRE-INITIAL CONTACT TO INITIAL CONTACT	**INITIAL CONTACT**	**INITIAL CONTACT TO 200 MS POST INITIAL CONTACT**
ANGULAR DISPLACEMENT	- ↓ FFTBA internal rotation angular displacement (affected: 10.04 ± 5.72, control: 13.93 ± 5.53, *p* = 0.046)	- No significant differences observed	- No significant differences observed
ANGULAR VELOCITY	- ↑ HFTBA eversion angular velocity (affected: -70.73 ± 35.41, control: -49.84 ± 24.86, *p* = 0.048) - ↑ knee adduction angular velocity (affected: 75.41 ± 28.51, control: 57.16 ± 21.55, *p* = 0.037) - ↑ trunk flexion velocity (affected: -62.01 ± 30.80, control: -40.63 ± 23.41, *p* = 0.025)	- ↑ HFTBA eversion angular velocity (affected: -44.19 ± 57.79, control: 16.24 ± 58.27, *p* = 0.004) - ↑ trunk flexion angular velocity (affected: - 57.30 ± 33.50, control: -31.20 ± 28.40, *p* = 0.017)	- ↑ knee flexion angular velocity (affected: -425.34 ± 48.30, control: -374.59 ± 56.94, *p* = 0.007) - ↑ knee external rotation angular velocity (affected: -67.37 ± 59.56, control: -64.60 ± 28.76, *p* = 0.048)
ANGULAR ACCELERATION	- ↑ peak negative hip frontal plane angular acceleration (affected: -1904.37 ± 1079.25, control: -1286.87 ±509.74, *p* = 0.035) - ↑ peak negative trunk sagittal plane angular acceleration (affected: -1430.11 ± 613.83, control: -992.85 ± 603.86, *p* = 0.038)	- ↑ peak negative knee sagittal plane angular acceleration (affected: -3450.92 ± 2048.42, control: -2087.70 ± 1095.33, *p* = 0.018)	- ↑ peak negative knee sagittal plane angular acceleration (affected: -3446.65 ± 1579.53, control: -2188.67 ± 1188.25, *p* = 0.011)

CAI Group Unaffected Limb Versus Matched Control

When comparing the unaffected limb of the CAI group to the healthy matched control no significant differences in angular displacement were observed during the pre-landing phase. A significantly decreased HFTBA external rotation velocity (Δ = 20.38°/s, p = 0.048) and increased peak trunk extension angular velocity (Δ = 12.53°/s, p = 0.011) in the unaffected limb of the CAI group when compared to the matched control during the pre-landing phase. A significantly increased peak negative hip transverse plane acceleration (Δ = 206.05°/s^2, p = 0.038) and increased negative trunk sagittal plane acceleration (Δ = 578.64°/s^2, p = 0.013) was also observed in the unaffected limb when compared to the healthy matched control during the pre-landing phase (Table 7.2). No other significant differences were observed between the unaffected limb and the matched control pre-landing (Appendix I, Table I.1, Table I.4, Table I.7).

At IC, a significantly decreased hip internal rotation angular velocity (Δ = 30.49°/s, p = 0.047) was observed in the CAI group unaffected limb when compared to a healthy matched control. A significantly increased peak negative trunk sagittal plane angular acceleration (Δ = 566.54°/s^2, p = 0.039) was also observed in the CAI group unaffected limb at IC (Table 7.2). No other significant differences were observed between the unaffected limb and the matched control at IC (Appendix I, Table I.2, Table I.5, Table I.8).

During the impact phase of landing decreased hip adduction angular displacement (Δ = 4.14°, p = 0.005) was observed in the unaffected limb of the CAI group when compared to the healthy matched control. Increased peak knee flexion angular velocity (Δ = 49.20°/s, p = 0.034), increased peak trunk flexion angular velocity (Δ = 30.14°/s, p = 0.003), decreased peak lateral flexion towards landing limb angular velocity (Δ = 48.18°/s, p = 0.009), increased peak trunk lateral flexion towards non-landing limb velocity (Δ =

48.51°/s, p = 0.002) and increased peak trunk external rotation angular velocity (Δ = 14.00°/s, p = 0.010) were observed in the unaffected limb of the CAI group during the impact phase of landing. Increased peak positive trunk sagittal plane angular acceleration (Δ = 369.39°/s^2, p = 0.011) was also observed in the unaffected limb during this phase (Table 7.2). No other significant differences were observed between the unaffected limb and the matched control during the post impact phase of landing (Appendix I, Table I.3, Table I.6, Table I.9).

Table 7.2 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the unaffected limb of the CAI group to a matched control.

UNAFFECTED COMPARED TO MATCHED CONTROL			
	200 MS PRE-INITIAL CONTACT TO INITIAL CONTACT	**INITIAL CONTACT**	**INITIAL CONTACT TO 200 MS POST INITIAL CONTACT**
ANGULAR DISPLACEMENT	- No significant differences observed	- No significant differences observed	- ↓ hip adduction angular displacement (unaffected: -0.17 ± 4.07, control: 4.31 ± 4.81, *p* = 0.005)
ANGULAR VELOCITY	- ↓ HFTBA external rotation angular velocity (unaffected: -49.98 ± 29.93, control: -70.36 ± 34.76, *p* = 0.048) - ↑ trunk extension angular velocity (unaffected: 32.49 ± 14.67, control: 19.96 ± 13.04, *p* = 0.011)	- ↓ hip internal rotation angular velocity (unaffected: 103.90 ± 49.57, control: 134.39 ± 38.65, *p* = 0.047)	- ↑ knee flexion angular velocity (unaffected:-418.13 ± 74.62, control: -368.93 ± 58.19, *p* = 0.034) - ↑ trunk flexion angular velocity (unaffected: -114.77 ± 26.70, control: -84.63 ± 29.50, *p* = 0.003) - ↓ trunk lateral flexion towards landing limb angular velocity (unaffected: 71.09 ± 57.63, control: 119.27 ± 45.32, *p* = 0.009) - ↑ trunk lateral flexion towards non-landing limb angular velocity (unaffected: -66.29 ± 53.51, control: -17.78 ± 18.01, *p* = 0.002) - ↑ trunk external rotation angular velocity (unaffected: -33.68 ± 19.11, control: -19.68 ± 9.65, *p* = 0.010)
ANGULAR ACCELERATION	- ↑ peak negative hip transverse plane angular acceleration (unaffected: -1696.98 ± 748.97, control: -1490.93 ± 758.46, *p* = 0.038) - ↑ peak negative trunk sagittal plane angular acceleration (unaffected: -1496.08 ± 814.68, control: -917.44 ± 463.71, *p* = 0.013)	- ↑ peak negative trunk sagittal plane angular acceleration (unaffected: -637.44 ± 904.71, control: -70.90 ± 662.24, *p* = 0.039)	- ↑ peak positive trunk sagittal plane angular acceleration (unaffected: 1822.69 ± 515.05, control: 1459.30 ± 519.16, *p* = 0.011)

CAI Affected Limb Versus CAI Unaffected Limb

When comparing the affected limb to the unaffected limb, a decreased peak positive trunk sagittal plane acceleration (Δ = 180.61°/s^2, p = 0.031) was observed in the affected limb of when compared to the unaffected limb during the pre-landing phase (Table 7.3). No other significant differences were observed between the affected limb and the unaffected limb pre-landing (Appendix I, Table I.1, Table I.4, Table I.7).

At IC, a significantly increased trunk flexion angular velocity (Δ = 9.79°/s, p = 0.048) was observed in the affected limb when compared to the unaffected limb of the CAI group (Table 7.3). No other significant differences were observed between the affected limb and the unaffected limb at IC (Appendix I, Table I.2, Table I.5, Table I.8)

During the impact phase of landing a significantly increased peak trunk lateral flexion towards landing limb angular velocity (Δ = 45.75°/s, p = 0.006), a significantly decreased peak trunk lateral flexion towards non-landing limb angular velocity (Δ = 51.16°/s, p = 0.001) and an increased peak trunk internal rotation angular velocity (Δ =18.87°/s, p = 0.041) in the unaffected limb of the CAI group (Table 7.3). No other significant differences were observed between the affected limb and the unaffected limb during the post impact phase of landing (Appendix I, Table I.3, Table I.6, Table I.9).

Table 7.3 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s²) when comparing the affected limb to the unaffected limb of the CAI group.

AFFECTED COMPARED TO UNAFFECTED LIMB			
	200 MS PRE-INITIAL CONTACT TO INITIAL CONTACT	**INITIAL CONTACT**	**INITIAL CONTACT TO 200 MS POST INITIAL CONTACT**
ANGULAR DISPLACEMENT	- No significant differences observed	- No significant differences observed	- No significant differences observed
ANGULAR VELOCITY	- No significant differences observed	- ↑ trunk flexion angular velocity (unaffected: - 47.51 ± 35.57, affected: - 57.30 ± 33.50, *p* = 0.048)	- ↑ trunk lateral flexion towards landing limb angular velocity (unaffected: 71.09 ± 57.63, affected: 116.84 ± 35.56, *p* = 0.006) - ↓ trunk lateral flexion towards non-landing limb angular velocity (unaffected: -66.29 ± 53.51, affected: -15.13 ± 16.54, *p* = 0.041) - ↑ trunk internal rotation angular velocity (unaffected: 40.94 ± 22.19, affected: 59.81 ± 33.81, *p* = 0.041)
ANGULAR ACCELERATION	- ↓ peak negative trunk sagittal plane angular acceleration (unaffected: 980.38 ± 472.68, affected: 799.77 ± 316.71, *p* = 0.031)	- No significant differences observed	- No significant differences observed

7.3.4 Discussion

The aim of this study was to explore the differences in discrete kinematic variables that may correlate to injury during single-leg landing between CAI participants' unaffected and affected ankles and to compare the same variables to a matched control group. As significant differences were observed in kinematics null hypothesis $H0_7$ can be rejected.

CAI group affected limb versus matched control

Several differences were observed between the CAI affected group when compared to the healthy matched control.

Prelanding- When investigating angular displacement, a decreased FFTBA internal rotation angular displacement was observed in the affected limb when compared to the healthy matched control during the pre-landing phase. Research suggests that internal rotation is often involved in the mechanism of lateral ankle sprains (Fong et al., 2009a; Fong et al., 2012; Konradsen & Voigt, 2002). It is possible this is a protective mechanism that may be displayed following previous ankle sprains to help to protect the ankle. Interestingly, this is the only significant difference between these groups in angular displacement, this suggests it may not be just the angle but the rate at which the angle is covered by a joint that may predispose those with ankle instability to recurrent injury. Also observed during the pre-landing phase was an increase in HFTBA eversion velocity again suggesting the group attempts to move into the more stable closed packed position more quickly as a protective response to the imminent landing. A significantly increased knee adduction velocity and a significantly increased hip frontal plane peak negative acceleration was observed in the affected limb, this increase in the rate of frontal plane motion suggests increased instability when compared to the matched control. Also in the pre-landing phase, a significantly increased trunk flexion velocity and significantly increased trunk sagittal plane peak negative acceleration was seen. This could be an attempt to lower the body's centre of gravity, although this was not measured in this study. The trunk accounts for approximately 35% of body mass (Kulas et al., 2008), therefore, fast flexion of this segment will lead to a large shift in momentum. If foot position is slightly altered or an unstable surface is present on landing, this momentum shift may place the ankle at increased risk of injury.

Initial contact- The increased HFTBA eversion velocity and increased trunk flexion velocity discussed during the pre-landing were also present at IC along with an increased sagittal plane peak negative acceleration. Again, this suggests that the affected limb group try to increase the rate at which they lower their centre of gravity and decrease their height away from the injured limb. The increase in knee flexion acceleration may be an attempt to try to decrease the impact forces through the lower extremity.

Impact phase- During the period from IC to 200 ms post contact an increased knee flexion velocity and increased knee sagittal plane peak negative acceleration was observed. This may suggest an attempt to rapidly lower the body's centre of mass or alternatively may display a lack of control with the movement.

The high number of differences in each phase observed in angular velocity and angular acceleration may suggest a lack of control within the injured group. Increased velocities have been linked to increased risk of injury. The differences observed in angular velocity and angular acceleration are in agreement with Williams et al. (2001) who suggested that the rate of motion rather than the motion itself is a critical factor in injury.

CAI group unaffected limb versus matched control

Comparisons were made between the unaffected limb of the CAI group to the healthy matched control limb using independent samples *t*-tests.

Prelanding- During the pre-landing phase of the single-leg land, a significantly decreased HFTBA external rotation velocity was observed. This suggests a decreased ability to move the joint out of a vulnerable position (Y. Li et al., 2018) potentially resulting in an increased risk of lateral ankle sprains to the unaffected limb. A significantly increased trunk sagittal plane peak negative acceleration again suggests an attempt to quickly modify the centre of gravity or a lack of control (Myers, Riemann, Hwang, Fu, & Lephart, 2003) in preparation for the imminent landing. A significantly increased hip transverse plane peak negative acceleration was also observed during the pre-landing phase this could potentially be a method to protect the contralateral limb from touch down in the event of an unstable landing.

Initial contact- At IC, a significantly decreased hip internal rotation velocity was observed. Although not measured within this study, it is possible this method is implemented to laterally shift the centre of gravity away from the affected limb to prevent touch down with the affected limb. Increased sagittal plane acceleration in the negative direction was also observed. It is thought this is done as previously mentioned to decrease the height of the centre of gravity and thereby increase stability.

Impact phase- The only significant difference observed in angular displacement between the unaffected limb and the healthy matched control was a decreased hip adduction displacement in the unaffected limb during the impact phase. It is suggested that this could be a method of protecting the affected limb from touch down by shifting the centre of gravity more laterally to aid the maintenance of balance to avoid the touch down with the affected limb. As observed between the affected limb and the healthy control limb, a significant increase in knee flexion velocity was also observed in the unaffected limb when compared to the healthy matched control. Again, it is thought that this may be to help absorb forces from impact and decrease the height of the centre of gravity. When analysing the trunk velocity, several significant differences were observed. Increased trunk flexion velocity may have been observed in an attempt to decrease the height of the body's centre of gravity. Decreased trunk lateral flexion (towards the landing limb) angular velocity, increased trunk lateral flexion (towards the non-landing limb) angular velocity, increased trunk external rotation velocity are thought to be an attempt to laterally shift the centre of gravity and maintain balance.

CAI group affected limb versus CAI group unaffected limb

The last comparison was between the affected and the unaffected limbs of the CAI group to determine whether any compensatory strategies were adopted.

Prelanding- During the pre-landing phase of the single-leg land a decreased peak negative trunk sagittal plane angular acceleration was observed in the affected limb when compared to the unaffected limb. However, it is unclear why this was observed.

Initial contact- Increased trunk flexion velocity was observed in the affected limb of the CAI group when compared to the unaffected limb. This difference was also found when comparing between the matched control group and the affected limb and could again be an attempt to lower the centre of mass more quickly.

Impact phase- During the impact phase of landing an increased trunk lateral flexion (towards the landing limb) and a decreased trunk lateral flexion (towards the non-landing limb) velocity was observed along with an increase in trunk internal rotation velocity. This could signify a decreased stability of the affected limb and thus an attempt to increase stability due to the rapid movement of the upper extremity.

Clinical implications

As with walking a decreased FFTBA internal rotation was observed prior to the initial contact in the affected limb when compared to the matched control limb ($\Delta = 3.89^{\circ}$). As this is a consistent finding across both movements this warrants further investigation as it remains unclear whether this is due to a restriction in joint range of motion or whether this is a modification in movement pattern. It also remains unclear whether this may be a protective strategy or one that predisposes the individual to the recurrent lateral ankle sprains.

Another key finding is the rapid changes in trunk motion with a significantly increased trunk flexion velocity ($\Delta = 21.38^{\circ}/s$) and increased peak negative trunk sagittal acceleration ($\Delta = 437.26^{\circ}/s^2$). With two thirds of the bodies mass being located above the hip (Konz et al., 2006), it is proposed that rapid trunk flexion may be a protective mechanism to lower the bodies' centre of gravity. Though it is also suggested that this comes with additional modifications in order to maintain centre of mass over the feet. Previous literature has observed a decreased soleus activation and an increased tibialis anterior activation to maintain postural control (Frank & Earl, 1990). Fast dynamic movements are also known to decrease the available time for neuromuscular corrections – delaying muscular recruitment and neural feedback (Granata & England, 2006). The Fitt law of motor control states that kinematic errors are increased with faster paced movements (Granata & England, 2006). Increased trunk flexion velocity was also observed between the affected limb and the matched control at initial contact but also when comparing the affected limb to the unaffected limb at initial contact in addition to this all five significantly different variables observed when comparing the affected and the unaffected limbs were observed in trunk kinematics. Thus, it is suggested this may be of interest to clinicians targeting preventative and rehabilitative strategies to patients.

Research investigating ankle sprain mechanisms as well as research investigating differences between healthy individuals and those with ankle instability has previously lacked detail referring to movement above the knee. The findings of this study suggest that when rehabilitating individuals who have suffered from a lateral ankle sprain care should be taken in order to address the whole kinetic chain rather than addressing the ankle in isolation. Equally, the current findings suggest that it may be also worth investigating angular velocities and angular accelerations during the rehabilitation process as it may not be the range that is the issue, rather the rate at which the range is covered that may be the cause or contributing factor of instability. When using a return to sport criteria concerning the satisfactory completion of movements (Chinn & Hertel, 2010), it may be necessary to also address the quality of these movements not only at the ankle but the entire kinetic chain.

Limitations

This research was solely investigating angular kinematics, however, future research should combine this with the use of a force plate to analyse angular kinetics in combination. It is also important to note that ankle instability was determined using a self-defined questionnaire, however, the questionnaire utilised has been extensively validated and was used in accordance with the IAC current recommendations. This may not discriminate appropriately against issues such as hypermobility.

Future research

Of the significant differences observed between groups within this study, only four were within foot kinematics with the remainder of the significant differences observed in the hip, knee, and trunk. These differences may be proximal adaptations due to damage sustained within the distal foot and ankle complex. Large magnitude movements such as those from the trunk, hip and knee may inadvertently place the foot/ankle complex at increased risk. To date, limited research exists analysing the role of the trunk, hip, and knee in the mechanism of ankle sprains suggesting the need to investigate the role this plays to determine whether a ground up approach or vice versa is implemented.

Future research should further investigate velocities and accelerations as this study suggests that these may play a more significant role than angular displacements. Lastly, a prospective study should be implemented to see whether findings of this study may be predictive of an ankle sprain and ankle instability risk.

7.3.5 Conclusion

When observing angular displacement, only one significant difference was observed in the affected limb when compared to the healthy matched control. This is in agreement with the previous walking study. The rest of the significant difference observed between groups were in angular velocity and angular acceleration suggesting the rate at which a range is covered may be a more prominent issue in ankle instability. Lastly, of the 30 significant differences observed between groups at the three investigated time points, only 4 were involving foot kinematics suggesting adaptations may be more prominent in the more proximal joints of the body in order to help increase stability.

7.3.6 Development of Research Within the Thesis

As with the walking study, the next phase of this section will be to investigate whether the significantly different variables observed during the landing task are able to predict the score on the IdFAI questionnaire. During the walking analysis, FFTBA transverse internal rotation displacement and FFTBA transverse peak positive acceleration prior to HS along with trunk lateral flexion displacement post HS were able to account for 49.2%

of the IdFAI score variance. Ideally, we would like the prediction value to be as high as possible therefore this will next be conducted during the landing manoeuvre.

7.4 Study 2 – Part 3 – Can significantly different kinematic variables observed during landing between individuals with and without ankle instability predict the Identification of Functional Ankle Instability Questionnaire Score?

7.4.1 Introduction

The previous chapter highlighted several differences that existed when comparing angular displacements, angular velocities, and angular accelerations during single-leg landing at 200 ms pre-IC, at IC and during the impact phase of landing.

In a research and clinical setting there is a reliance on self-reported questionnaires for the classification of ankle instability (Simon et al., 2014). The Identification of ankle instability questionnaire uses a cut off score of 11 or more to classify instability (Gribble et al., 2013). Subsequently, an individual scoring 10 on this questionnaire would be deemed 'healthy'. This study will aim to identify whether any of the already observed differences between groups could be used to predict the score on the IdFAI questionnaire.

This study aims to identify whether we can predict scores on the IdFAI questionnaire from the significantly different variables observed during walking.

This study will address the following hypothesis previously presented in Chapter 2.19:

- **H_8** - Significant differences observed during single-leg landing will be able to predict IdFAI questionnaire score

7.4.2 Methods

Variables where significant differences were observed between the affected limb of the CAI group and the healthy matched control limb during single-leg landing (study 2, part 2 (7.3.3)), were used for analysis. These are outlined in Table 7.4.

Table 7.4 Variables with significant differences in study 2 part 2 used within the regression analysis.

	200 ms pre- initial contact	Initial contact	Initial contact – 200 ms post-initial contact
Angular displacement	- ↓ FFTBA internal rotation		
Angular velocity	- ↑ HFTBA eversion - ↑ knee adduction - ↑ trunk flexion	- ↑ HFTBA eversion - ↑ trunk flexion	- ↑ knee flexion - ↑ knee external rotation
Angular acceleration	- ↑ peak negative hip frontal plane - ↑ peak negative trunk sagittal plane	- ↑ peak negative knee sagittal plane	- ↑ peak negative knee sagittal plane

Statistical Analysis

Normality was assessed using the Shapiro-Wilks test. Due to the sample size used, Pearson's correlation analysis were performed to identify whether relationships between IdFAI questionnaire score and kinematic variables existed. All predictors that showed a moderate (r = 0.3 upwards; Cohen & Cohen, 1977) and significant correlation ($p < .05$) were kept for further analysis, while all other variables were removed from subsequent analysis. Linear or multiple stepwise regressions were then used to identify which kinematic variables during single-leg landing best predicted IdFAI score. Data were inspected to ensure it met all assumptions of a regression before proceeding as outlined in 6.4.2. Data is presented as group means and standard deviations and the symbol Δ denotes differences between group means.

7.4.3 Results

Significant correlations can be observed in Table 7.5. Of the 12 variables included in the analysis for this study, 5 variables displayed a moderate and significant correlation and were inputted into a stepwise regression analysis.

The stepwise regression analysis showed knee sagittal plane peak negative acceleration from IC to 200 ms post-IC (impact phase) to be the best independent predictor of IdFAI score ($R = 0.520$, $R^2 = 0.270$, $F = 12.576$, $p = 0.001$). Combined knee sagittal plane peak negative acceleration from IC to 200 ms post-IC and knee frontal adduction velocity from 200 ms prior to IC to IC improved the prediction ($R = 0.593$, $R^2 = 0.352$, $F = 4.152$, $p = 0.050$). Prediction equations are outlined in Table 7.6 Beta and standard error values are outlined in Table 7.7.

Table 7.5 Pearson's correlation outputs for kinematic variables compared to IdFAI score

		FFTBA transverse internal rotation displacement 200-IC	HFTBA frontal Eversion velocity 200-IC	Knee frontal adduction velocity 200-IC	Trunk sagittal FLEX velocity 200-IC	Hip frontal min acceleration 200-IC	Trunk sagittal min acceleration 200-IC	HFTBA frontal Eversion velocity ICVALUE	Trunk sagittal FLEX velocity ICVALUE	Knee sagittal min acceleration ICVALUE	Knee sagittal FLEX velocity IC-200	Knee transverse ER velocity IC-200	Knee sagittal min acceleration IC-200
IdFAI SCORE	Pearson Correlation	-0.3103	-0.295	0.492*	-0.2402	-0.1674	-0.1871	-0.411*	-0.256	-0.497*	-0.378*	0.046	-0.520*
	Sig. (2-tailed)	0.065	0.080	0.002	0.158	0.329	0.275	0.013	0.132	0.002	0.023	0.790	0.001

Table 7.6 Prediction equations for IdFAI score

Model	Equation
Model 1	IdFAI Score = (-0.181 x Knee sagittal min acceleration IC-200) + 2.415 (± 8.268)
Model 2	IdFAI Score = (-0.130 x Knee sagittal min acceleration IC-200) + (0.115 x Knee frontal adduction velocity 200-IC) - 2.568 (± 7.909)

Table 7.7 Unstandardized and standardized Beta values for each of the 9 regression models

Dependent variable	Variable	B	SE	*ß*	st error of estimate
IdFAI Score (model 1)	Constant	2.415	2.985		
	Knee sagittal min acceleration IC-200	-0.181	0.051	-0.520	8.268
IdFAI Score (model 2)	Constant	-2.568	3.760		
	Knee sagittal min acceleration IC-200	-0.130	0.055	-0.372	
	Knee frontal adduction velocity 200-IC	0.115	0.057	0.321	7.909

7.4.4 Discussion

Completion of the IdFAI questionnaire places individuals on a scale on ankle instability with higher scores suggesting increased instability and scores lower than 10 suggesting ‘healthy’ individuals. It is suggested that this score may be used as a continuum and as such have investigated whether significantly different variables during single-leg landing may be used to predict the ‘severity’ of this instability.

Of the 12 variables included within this analysis 5 were significantly correlated to the IdFAI questionnaire score. The regression analysis produced 2 models. The first used knee sagittal plane peak negative acceleration from IC to 200 ms post-IC. This value may signify either a sudden halt in knee flexion or a faster increase in knee flexion when compared to the healthy control group possibly in an attempt to rapidly lower the centre of gravity or alternatively demonstrating a lack of control. The knee acceleration alone is able to predict 27% of the variance observed. This combined with the increased knee frontal plane adduction velocity prior to IC increases the prediction to 35.2%. As models were able to help predict the IdFAI questionnaire score null hypothesis $H0_8$ can be rejected.

Clinical implications

More research is needed to confirm whether the above models may be used to strengthen the categorization of instability. The number of variables that correlate to the IdFAI score may also suggest that the cut off value used to group instability may be better interpreted on a scale. Further prospective research is needed to determine whether these factors predisposed individuals to the initial ankle injury or whether these are subsequent adaptations that have occurred as a result of the initial injury.

7.4.5 Conclusion

Knee sagittal plane peak negative acceleration from IC to 200 ms post-IC combined with knee frontal plane adduction velocity from IC to 200 ms post-IC were able to predict 35.2% of the IdFAI score variance. It would be key to investigate other movements such as a cutting manoeuvre which are also commonly associated with ankle sprains to see if this percentage can be increased.

7.4.6 Development of Research Within the Thesis

Previous research during walking identified FFTBA transverse internal rotation displacement and FFTBA transverse peak positive acceleration prior to IC along with trunk lateral flexion displacement post-HS were able to account for 49.2% of the IdFAI score variance. This model does not identify any of the same variables for use within the regression model. The percentage the single-leg land was able to predict was only 35.2% in the present study in comparison to the 49.2% observed in the walking study. The next chapter will investigate a cutting manoeuvre a common injury mechanism for a lateral ankle sprain. It may also be beneficial to combine each of the variables obtained in the walk, land and cutting manoeuvre to see if a stronger prediction is available, though, this may need increased numbers to improve the power of this.

Chapter 8.0 Study 3 – Analysis of Cutting Manoeuvre Muscle Activation and Kinematics in Individuals with Ankle Instability and Healthy Control Participants

8.1 Cutting Manoeuvres Chapter Overview

Lateral ankle sprains commonly occur when the centre of gravity is shifted laterally over the lateral border of the weight bearing foot causing high-velocity inversion to occur (Dubin et al., 2011). Sports requiring sudden cutting manoeuvres, jumping, landing and sudden stops such as basketball and football have been found to account for the highest percentages of ankle sprain as the athlete is at risk of catching the lateral edge of the foot or landing on another player's foot and causing the ankle to roll into combined inversion and plantarflexion (Knight & Weimar, 2011b; McGuine & Keene, 2006). Bahr and Krosshaug (2005) stated that in order to formulate an effective preventative measure, a thorough understanding of the injury mechanism is needed. It is thought that quantifying joint kinematics in individuals with ankle instability may help to understand why these individuals experience the 'giving way' sensation and recurrent ankle sprains. This understanding can then be used to aid the development of rehabilitation and preventative or screening measures (Koshino et al., 2014).

An audit of injuries in football in the English Premier League found 39% of ankle sprains to occur in non-contact situations. Twisting and turning accounted for 21% of these injuries (Woods et al., 2003). Bloomfield, Polman and O'Donoghue (2007) found a majority of turns were between 0 and 90° and performed approximately 700 times per match by defenders, 600 times by strikers and 500 times by midfielders. Previous biomechanical analysis has focused on smaller cutting angles (e.g. 45°) (Sigward, Cesar, & Havens, 2015) as such it is felt pertinent to investigate cutting manoeuvres in individuals who suffer recurrent ankle sprains during larger cutting angles often seen during sports.

Numerous studies have investigated the impact of preventative measures on kinematics and muscle activation during cutting manoeuvres (Gribble, Radel, & Armstrong, 2006; Gudibanda & Wang, 2005; W. C. Lee, Kobayashi, Choy, & Leung, 2012). Very few researchers first investigate the differences in kinematics and activation in individuals with ankle instability and those without during cutting manoeuvres to highlight the areas that need to be addressed. More thorough understanding of biomechanics will enable more effective intervention and prevention strategies to be adopted.

As established in previous studies, modelling of the foot as one rigid segment excludes motion between different segments of the foot providing inadequate information on the biomechanics (Stebbins et al., 2006). De Ridder et al. has previously documented multi-segmental foot motion during walking (De Ridder et al., 2013) and landing (De Ridder et al., 2015b) concluding that the multi-segmental foot model provided greater details of the intricacies of the foot. To the author's knowledge, no research has utilised the multi-segmental model during cutting manoeuvres, a common mechanism of ankle sprains. As such it is felt that this may provide further details of the movement between segments during this manoeuvre and combined with a full body marker set may give valuable information regarding the movement mechanics of those with CAI which may lead to improved intervention strategies.

As with Chapter 6.0 and Chapter 7.0, this chapter will adopt a 3-part structure to further analyse differences in muscle activation and movement kinematics. Chapter 8.2 will analyse full time series movement patterns and muscle activation patterns between individuals with CAI and healthy controls during a cutting manoeuvre. Chapter 8.3 will address the angular displacements, angular velocities and angular accelerations displayed by groups during a cut. Lastly, Chapter 8.4 will view whether the findings observed in Chapter 8.3 can be used as a prediction of the individuals' IdFAI questionnaire score.

8.2 Study 3 – Part 1 - Full Gait Cycle Analysis of Lower Limb and Trunk Kinematics and Muscle Activations During a Cutting Manoeuvre in Participants with and Without Ankle Instability

8.2.1 Introduction

Analysis of movement patterns using full time series analysis has proven to be beneficial for analysis of walking (De Ridder et al., 2013) and landing (De Ridder et al., 2015a; De Ridder et al., 2015b) however to the author's knowledge this has not been conducted in cutting movements, a mechanism commonly associated with lateral ankle sprains (Knight & Weimar, 2011b; McGuine & Keene, 2006).

This study will again utilise statistical parametric mapping to better identify differences in movement and muscle activation patterns at exact periods of the cutting manoeuvre.

Activation of muscles prior to and in response to joint loading and motion has previously been postulated to be a major factor in joint stability (Riemann & Lephart, 2002). Previous research has investigated ankle, knee and hip kinematics and muscle activation during a land and cut manoeuvre and observed differences in muscle activation patterns between affected and control groups (Son et al., 2017), however, the isometric squat was used as a reference task. Whilst populations prone to injury may have difficulty producing a maximal voluntary isometric contraction (Bolgla & Uhl, 2007; Konrad, 2005) an isometric squat is not a comparable movement and participants may adopt different strategies for this movement thus further research may be necessary to confirm these findings.

This study will combine analysis of the trunk, hip, knee and multi-segmental foot kinematics and sEMG activation patterns during the period prior to and the period following IC during a cutting manoeuvre. As ankle sprains often occur during cutting manoeuvres it is suggested that this exploratory study of movement pattern will provide increased knowledge to clinicians for use when developing preventative and rehabilitation strategies for ankle sprains. The aim of this study was to compare trunk, hip, knee and multi-segmental foot kinematics and muscle activation during cutting between participants with CAI and healthy controls.

This study will address the following hypotheses previously presented in Chapter 2.19:

- **H_9** - CAI participants will display modified kinematic movement patterns (SPM) during cutting
- **H_{10}** - CAI participants will display modified muscle activation patterns (SPM) during cutting

8.2.2 Methods

Participants

Eighteen healthy controls (age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 participants with CAI (13 males, 5 females, age: $\bar{\chi}$ = 22.0 ± 2.7 years, height: $\bar{\chi}$ = 176.8 ± 7.9 cm, mass: $\bar{\chi}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were included in this study as outlined in General Methods 3.1.

Protocol

Participants performed a 5-minute warm-up on a cycle ergometer at 60 Watts. Bilateral electromyographic (EMG) data were recorded for the gluteus medius, peroneus longus and tibialis anterior during the cutting manoeuvres as outlined in general methods

Chapter 3.4. EMG data were normalised to the activity mean for each cut (Bolgla & Uhl, 2007). Motion analysis data were recorded using an Owl Digital Real Time 10 camera system (Motion Analysis, Santa Rosa, California) sampling at 200 Hz using the Helen Hayes marker set (Davis et al., 1991) combined with the Oxford foot model (Stebbins et al., 2006; Wright et al., 2011) general methods Chapter 3.5.

Participants were required to perform three cutting manoeuvres on their left and right limbs. The trial limb was randomised to minimise the effect of fatigue. Participants were instructed to stand with feet shoulder width apart with weight equally distributed over both feet on a 30 cm box (Kunugi et al., 2017). They were instructed to jump two-footed forward off the box landing two-footed before performing a 90° cut (Figure 8.1). The jump to cut manoeuvres was chosen to replicate sporting activity and to reduce deviation in approach speed. One-minute rest was provided between trials. Trials were discarded if errors were observed in marker tracking. Each participant performed a familiarisation of the movement until they were comfortable with the movement before recording.

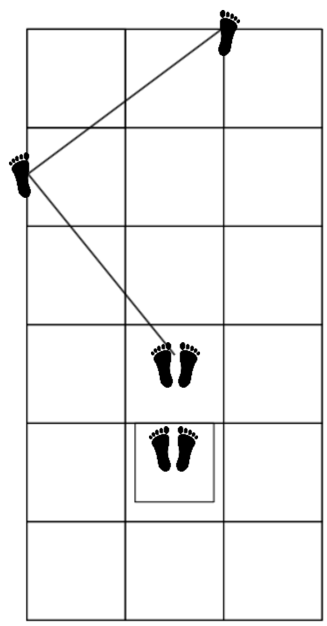


Figure 8.1 90 degree cutting manoeuvre

Data Reduction and Statistical Analysis:

Data were inspected using Cortex (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa) software before importing into Visual 3D (Visual3D v6 x64, C-motion, Germantown) for analysis. Data were smoothed using a 6Hz Butterworth filter. Initial contact was determined using the method proposed by O'Connor, Thorpe, O'Malley and Vaughan (2007). Following inspection of several methods, this was deemed to be the most accurate for the movement under analysis. This method creates a new signal by calculating the midpoint between the heel (CAL1) and toe marker. The first derivative is then calculated on the vertical component of the signal. Event markers were created at the minimum value for IC. Toe off was determined using peak knee extension angle (Dingwell et al., 2001; Fellin et al., 2010). EMG data were root mean squared by a moving window of 100 ms and normalised to mean task activation. Kinematic and EMG data were exported for two time durations- 200 ms pre-IC to IC and from IC to toe-off into tab-delimited text files for analysis using Matlab.

Cutting velocity was recorded by observing the pelvis segment peak and average angular velocity. Independent samples *t*-tests were performed to check for differences between the CAI group and the control group.

Kinematic data were exported for forefoot-hindfoot angle, forefoot-tibia angle, hindfoot-tibia, cutting hip, knee, trunk and non-cutting hip in all planes of motion. So as not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment (De Ridder et al., 2013; Wright et al., 2011). Data were analysed using SPM. Analysis was performed in Matlab 2016a (The MathWorks, Natick, USA) using the SPM1D open-source package (spm1d.org). Data were tested for normality using a D'Agostino-Pearson's test. A matched control limb was compared to the CAI groups affected limb and unaffected limbs using an independent samples *t*-test (α = 0.05). The unaffected and affected limb of the CAI group were then compared using a paired samples *t*-test (α = 0.05).

8.2.3 Results

Independent-samples *t*-tests revealed no significant differences ($p > 0.05$) between groups for age, height, mass, or shoe size. Independent samples *t*-test also revealed no significant differences between the control group and CAI group groups for average (CAI = 1.30 ± 0.17 $m.s^{-1}$, control = 1.31 ± 0.12 $m.s^{-1}$) and maximum (CAI = 1.63 ± 0.21 $m.s^{-1}$, control = 1.65 ± 0.17 $m.s^{-1}$) cutting velocity.

Matched Control vs Affected Limb

During the pre-contact period a significantly decreased forefoot-tibia internal rotation was observed in the affected limb when compared to the matched control at 0-11% (mean difference = 4.98°, peak difference = 5.40°, $p = 0.048$, Figure 8.2). No significant differences were observed in this variable during the 200 ms following IC (0, Figure K.8). No significant differences were observed in FFHFA (0, Figure K.1, Figure K.7), HFTBA (0, Figure K.2, Figure K.9), hip (0, Figure K.3, Figure K.10), knee (0, Figure K.4, Figure K.11) or trunk (0, Figure K.5, Figure K.12) kinematics or in gluteus medius, tibialis anterior and peroneus longus muscle activation patterns (0, Figure K.6, Figure K.13) during the pre-contact period or the 200 ms following IC between the matched control and the affected limb.

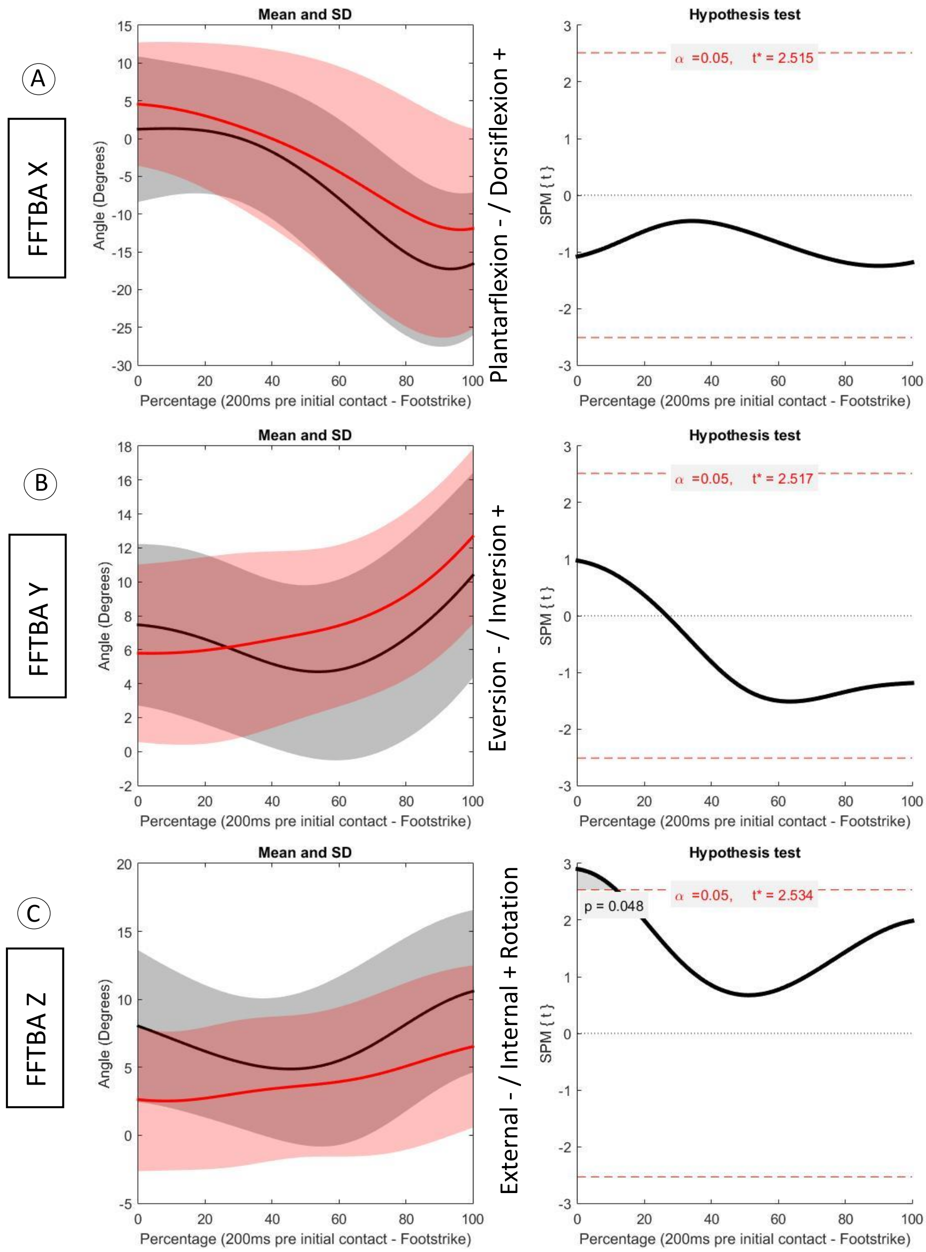


Figure 8.2 Forefoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

Unaffected vs Affected Limb

During the pre-contact phase of the cut a significant difference in FFTBA frontal plane motion was observed when comparing the unaffected limb to the affected limb at 56-73% (mean difference=3.71°, peak difference=3.76°, p = 0.045, Figure 8.3). No significant differences were observed in FFTBA during the 200 ms post IC (0, Figure K.21).

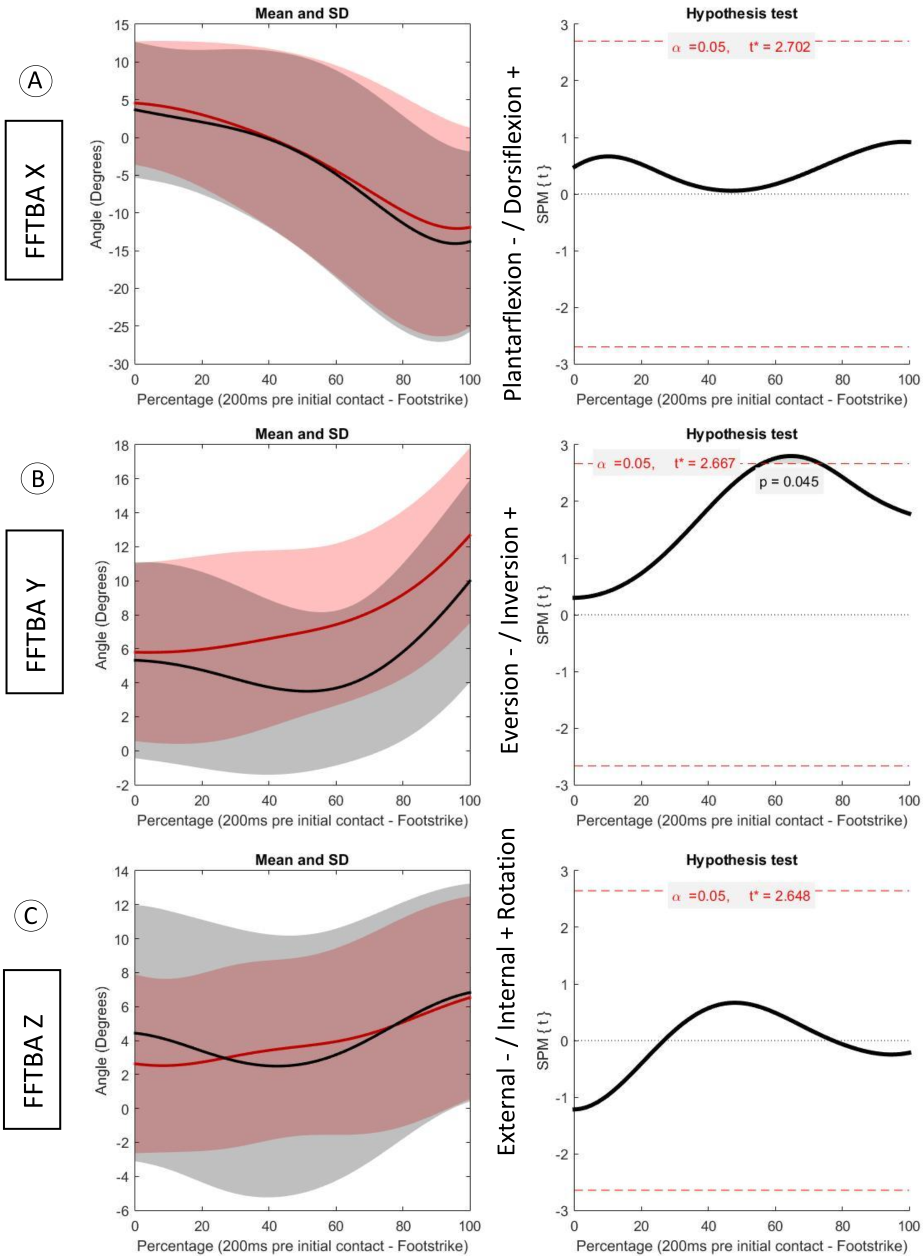


Figure 8.3 Forefoot – Tibia angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected -— Affected - —) and t-test output.

No significant differences were observed in FFHFA (0, Figure K.14, Figure K.20), HFTBA (0, Figure K.15, Figure K.22), hip (0, Figure K.16, Figure K.23), knee (0, Figure K.17, Figure K.24) or trunk (0, Figure K.18, Figure K.25) kinematics or in gluteus medius, tibialis anterior and peroneus longus muscle activation patterns (0, Figure K.19, Figure K.26) during the pre-contact period or the 200 ms following IC between the matched control and the affected limb.

Matched Control vs Unaffected Limb

When comparing the unaffected limb of the ankle instability group to the matched control prior to IC a significant difference was observed in FFTBA frontal plane motion at 68-90% (mean difference=5.28°, peak difference=5.42°, p = 0.044, Figure 8.4). No other significant differences were observed in FFTBA kinematics (0, Figure K.33).

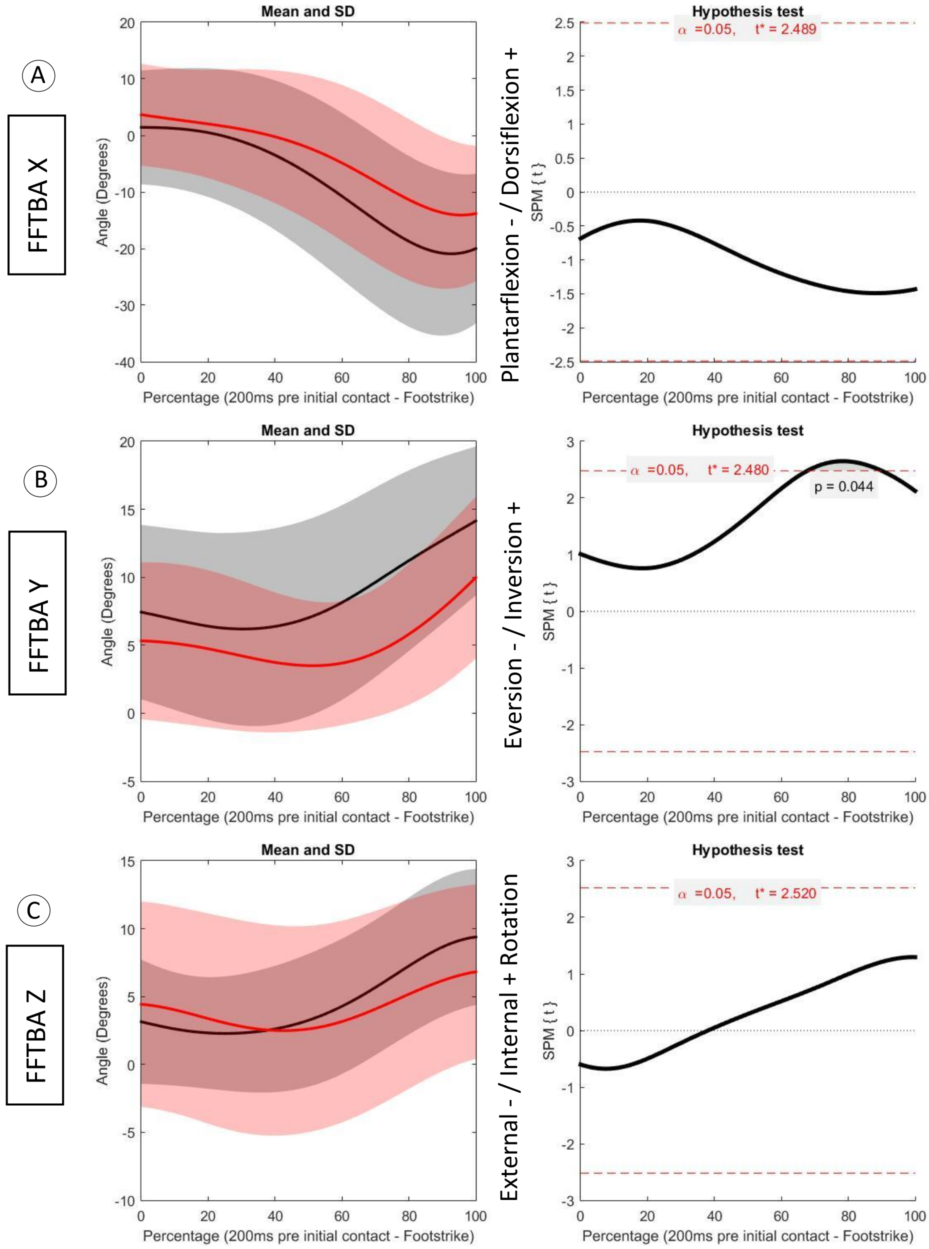


Figure 8.4 Forefoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control – — Unaffected - —) and *t*-test output.

No other significant differences were observed in kinematics between the groups (0, Figure K.27-Figure K.37). When investigating patterns of muscle activation a significantly increased peroneus longus activation was observed in the unaffected limb at 0-20% of the pre-contact phase of the cut (mean difference=36.04%, peak difference=38.81%, p = 0.014, Figure 8.5).

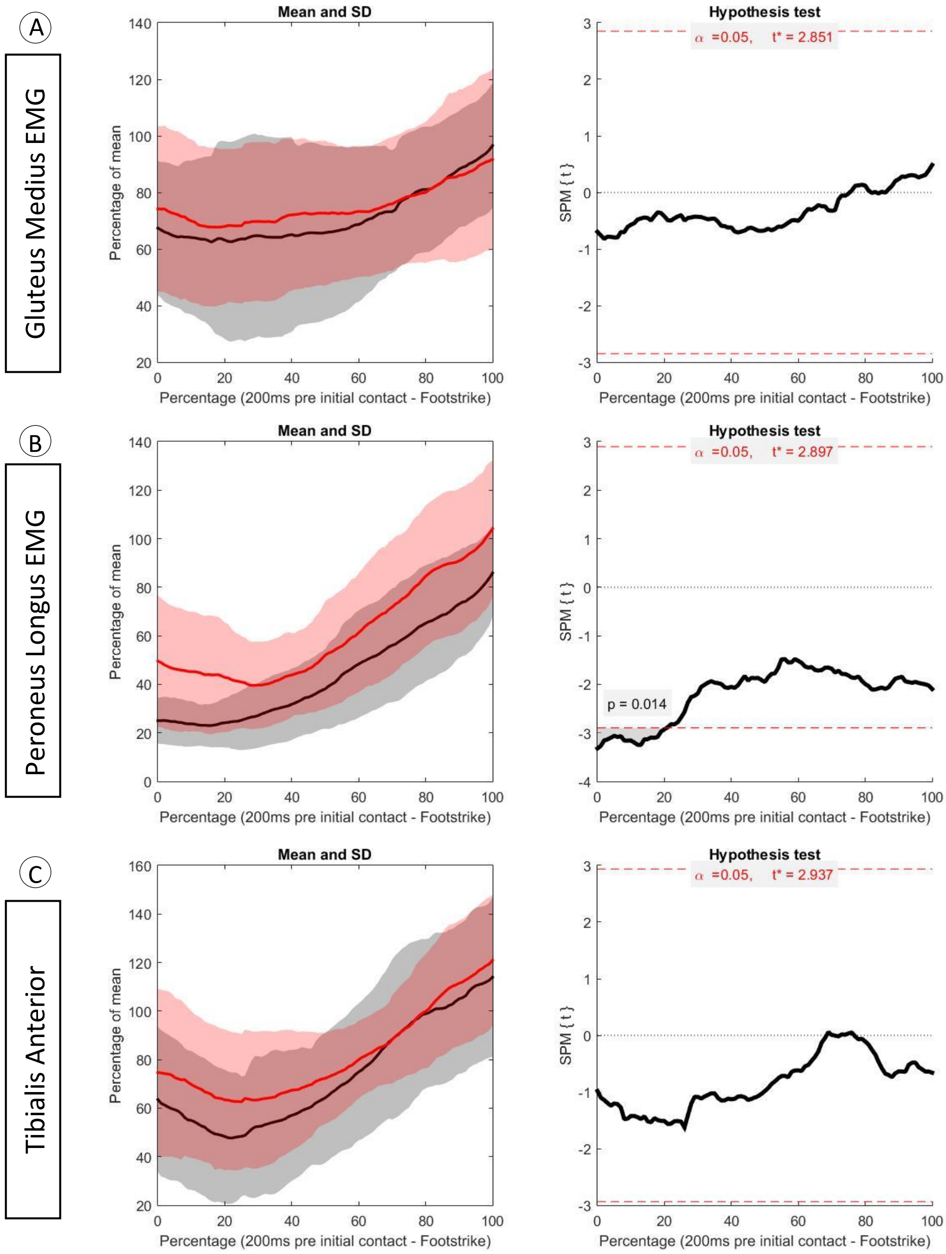


Figure 8.5 Gluteus Medius and Tibialis Anterior muscle activation - 200 ms pre IC to IC - means and standard deviations (Matched Control -— Unaffected - —) and *t*-test output.

Significantly decreased peroneus longus activation was observed in the unaffected limb from 49-64% of the IC to toe-off phase of the cut (mean difference= 64.50%, peak difference=71.38%, p = 0.001, Figure 8.6). No other significant differences were observed in muscle activation patterns.

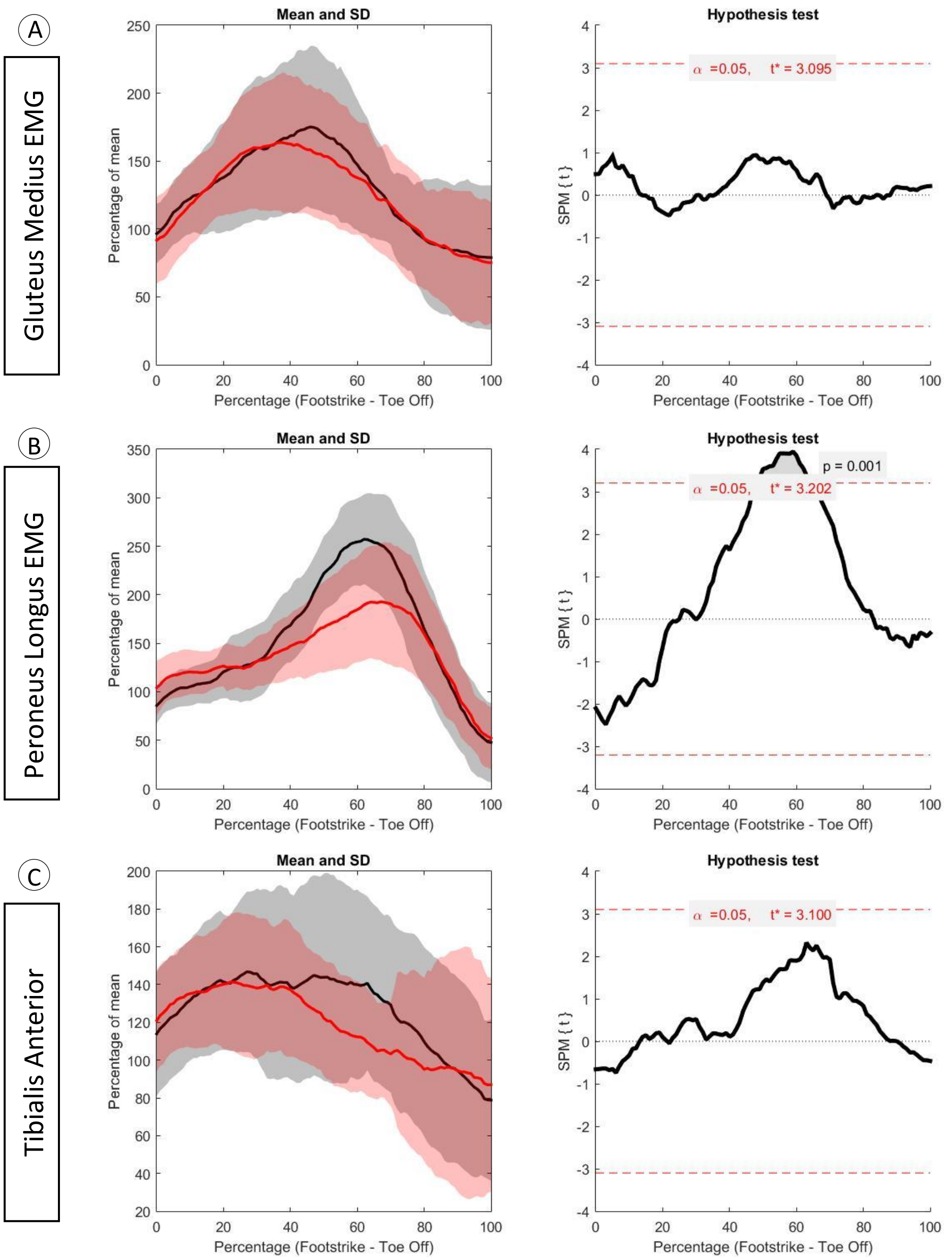


Figure 8.6 Gluteus Medius and Tibialis Anterior muscle activation – IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

8.2.4 Discussion

This study aimed to analyse the differences in kinematic and muscle activation patterns between individuals with and without ankle instability during a cutting manoeuvre (a common mechanism for ankle sprains). As significant differences were observed in kinematic and electromyographic movement patterns both null hypothesis $H0_9$ and $H0_{10}$ can be rejected.

Decreased FFTBA internal rotation

When analysing the differences between the matched control and the affected limb a significantly decreased FFTBA internal rotation was observed at 0-11% and although not significant the mean for FFTBA internal rotation remained lower in the affected limb of the instability group when compared to the matched control. Internal rotation is increasingly suggested as a mechanism for ATFL damage and as such this position may be an attempt of the limb to increase stiffness to protect the joint from further damage. The authors propose that this reduction in FFTBA internal rotation angular displacement prior to IC may in fact place individuals at an increased risk of ankle sprains. It is postulated that the small degree of internal rotation displayed by individuals may place the ankle in a more closed packed position allowing the talus and joint capsule to restrict frontal plane motion. It has previously been postulated that an increased medial deviation of the subtalar joint axis results in increased pronation moments (Kirby, 2001). Prospective research is needed into this variable to determine whether this is something that may predispose individuals to the initial ankle sprain or whether this modification may be because of the sprain. Further research should investigate whether internal rotation is blocked, potentially due to differences in bone shape or positioning that have been documented in those with ankle instability (Tümer et al., 2019) or whether this is an active modification.

As no measurements of participant's anatomical range of motion were taken prior to the trial it is unclear whether participants were limited with internal rotation due to anatomical variations or previous injury. Some research has also suggested a decreased ROM in ankles with CAI (Hoch, Staton, Medina McKeon, Mattacola, & McKeon, 2012). Whilst other research has observed anterior positional faults of the talus (Wikstrom & Hubbard, 2010) and fibula (Hubbard, Hertel, & Sherbondy, 2006) following ankle sprains. This is the first study to observe this finding and as such requires further investigation. It is also unclear whether this change in movement kinematics is because of the injury or was present prior to the initial ankle sprain. No other significant differences were observed in kinematic movement patterns or in muscle activation patterns between groups.

Forefoot-tibia inversion/eversion

A significant difference was observed in FFTBA frontal plane motion between the unaffected and the affected limb (at 56-73% of the pre-contact phase) and when comparing the matched control and the unaffected limb (at 68-90% of the pre-contact phase). Closer analysis would suggest that the unaffected limb displays less inversion than the affected limb and the healthy matched control. This outlines that compensatory strategies are adopted by the unaffected limb possibly to better 'brace' the limb in anticipation of the next step onto the affected limb. There is a paucity of research investigating the contralateral limb of those with CAI so this may highlight a key area for future focus. Previous studies also observed significant differences in frontal plane motion of the FFTBA during HS to toe off during walking (Northeast et al., 2018).

Interestingly minimal research has investigated the unaffected limb however this finding may suggest more focus should be given to ensure symmetrical movement patterns are adopted by individuals as part of the rehabilitation protocol. No further differences were observed between groups for kinematics and muscle activation patterns. Interestingly no significant differences were observed in hip and knee kinematic movement patterns. This is contrary to the results of Koshino et al. (2016) and Son et al. (2017) who observed increased hip flexion during the stance phase. Son et al. (2017) increased knee flexion and abduction during the stance phase along with less hip abduction. These findings were similar to Koshino et al. (2014). These differing findings may be due to methodological differences in the manoeuvre performed or differences in the method of analysis.

Electromyography

Electromyography comparing the unaffected limb of the instability group to the matched control observed an increase in peroneus longus muscle activation patterns when normalised to mean activity at 0-20% of the pre-contact stage. This may suggest pre-activation as a feedforward strategy in anticipation of IC. This is contrasting to the research of Delahunt et al (2006b) who observed a decreased peroneus longus muscle activation prior to IC during a single leg drop jump. These findings are also similar to that of Caulfield et al. (2004) who also observed decreased peroneus longus EMG prior to IC during landing. The prior findings were observed in the affected limb when compared to a healthy control. This may explain the discrepancy in observations. It may also suggest a compensatory strategy again adopted by the unaffected limb to protect the affected limb. Interestingly peroneal muscle activation patterns significantly decreased during the period from 49-64% of the IC to toe off phase of the cut potentially suggesting a lack of control following IC.

A limitation of this study is that electromyography was compared to mean activity activation due to the difficulty in obtaining a maximal voluntary isometric contraction in the injured limb. This makes it difficult to compare muscle activation and instead means it is necessary to look purely at the patterns of activation. This research uses time series analysis which is beneficial for investigating the patterns of movement however this method involves time normalisation and as such may not provide information on the speed at which the movement was performed. Therefore, subsequent analysis is necessary into the velocities and acceleration of the kinematics reported in this study. This will provide further information as previous research suggests it may be the speed at which displacement is covered and not the displacement itself that may prove to be an issue for injured athletes.

Clinical implications

This study highlighted 5 key differences in kinematic and muscle activation patterns between groups. When comparing the matched control to the affected limb a decreased FFTBA internal rotation was observed in the affected limb at 0-11% pre initial contact (mean difference 4.98°). Previous cadaveric studies have observed maximal internal rotation range of 17.62° (Wilkerson et al., 2010), thus a mean difference of 4.98° may have a clinically significant impact on movement adopted. Further research is needed to determine whether this is a modification of movement subsequent to the initial sprain or a movement pattern already adopted prior to an initial sprain. It is also necessary to assess whether this is due to a lack of range at the joint itself. Differences in FFTBA frontal plane motion were observed when comparing the matched control and the unaffected limb (mean difference = 5.28°) and when comparing the unaffected and the affected limb of the CAI group (mean difference = 3.71°). Overall frontal plane range of motion has been previously reported to be 35° (Brockett & Chapman, 2016) thus a difference of 5.28 and 3.71 degrees respectively is clinically important. It appears that the unaffected limb of the CAI group appears to adopt a less inverted position. A fourfold increase in injury risk was observed during a competition season in triathletes with a supinated foot posture (Burns, Keenan, & Redmond, 2005), suggesting the modification of the unaffected limb may increase injury risk. The authors would suggest that interventions targeting forefoot-tibia kinematics in both the unaffected and affected limbs may prove to be beneficial.

8.2.5 Conclusion

This study identified a decreased FFTBA internal rotation in the affected limb of the ankle instability group when compared to a healthy matched control which it is suggested may be a protective mechanism but potentially also predisposes individuals with instability to recurrent ankle sprains. This study also highlights the need for bilateral rehabilitation programmes and prevention strategies for those suffering from recurrent sprains exposing differences present in the unaffected limb. Further research should identify the velocity at which the movements identified in this study are covered to identify further risk factors of instability.

8.2.6 Development of Research

Study 1 part 1 (Chapter 6.2) also observed significant differences in FFTBA frontal plane kinematics between the affected limb and the unaffected limb of the CAI group however they observed this during the stance phase as opposed to the pre-contact phase. This highlights that different movement strategies are adopted between limbs and thus careful bilateral movement analysis and rehabilitation is important. SPM allows a useful analysis of movement and muscle patterns however it does require data to be time normalised. The next study will analyse differences in peak angular displacements and the angular velocity and angular acceleration to observe whether although similar movement patterns are adopted, (except the observed significant difference in FFBTBA internal rotation) between the affected limb and the healthy matched control, whether the speed at which these movements occur may be impacting their predisposition to ankle sprains.

8.3 Study 3 – Part 2 - A Comparison of Lower Limb Angular Displacements, Angular Velocities and Angular Accelerations During a Cutting Manoeuvre Between Participants with Chronic Ankle Instability and Healthy Controls

8.3.1 Introduction

As with walking and single-leg landing, angular displacements are often reported when reporting kinematics during cutting mechanisms (Koshino et al., 2014; Koshino et al., 2016; Son et al., 2017). Increased angular velocity has previously been observed in research on a tilt platform (Vaes et al., 2001). This significantly increases stress on ligaments which may lead to a sprain. Minimal research currently exists investigating angular velocity and even less angular acceleration during cutting.

Accidental ankle sprains have been analysed within cutting manoeuvres and studies report increases in angular velocities (Kristianslund et al., 2011; Mok et al., 2011). Kristianslund et al. (2011) asked participants to side cut around a static defender during which a participant experienced a lateral ankle sprain. They reported a significantly increased inversion velocity at 559 ˚/s in the sprain trial vs 166 ˚/s and 221 ˚/s in the previous control trials. Similarly, in the Beijing Olympic Games, two ankle sprains were analysed from televised video recordings the second was sustained during a field hockey match whilst the player was running under pressure and inversion velocity was reported as 1397 °/s calculated by using a skeletal matching approach and then calculating in accordance with the joint coordinate system (Mok et al., 2011). These findings led the authors to question whether displacement data alone is sufficient to highlight potential differences between individuals with CAI and healthy controls.

The aim of this study was to compare angular displacement, velocities and accelerations of the trunk and lower limb from the sagittal, frontal and transverse planes during the pre-contact, contact and post-contact phases of a lateral cutting manoeuvre between participants with CAI and healthy controls.

This study will address the following hypothesis previously presented in Chapter 2.19:

- **H_{11}** - CAI participants will display modified discrete kinematic variables during cutting

8.3.2 Methods

Participants

Eighteen healthy controls (14 males, 4 females, age: $\bar{\chi}$ = 22.4 ± 3.6 years, height: $\bar{\chi}$ = 177.8 ± 7.6 cm, mass: $\bar{\chi}$ = 70.4 ± 11.9 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) and 18 participants with CAI (13 males, 5 females, age: $\bar{\chi}$ = 22.0 ± 2.7 years, height: $\bar{\chi}$ = 176.8 ± 7.9 cm, mass: $\bar{\chi}$ = 74.1 ± 9.6 kg, UK shoe size *Mdn*: 8.5, MIN: 4, MAX: 11) were included in this study as outlined in General Methods 3.1.

Protocol

Kinematic data collected and imported into visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland, USA) as in Chapter 8.2 were further analysed in this section of the study.

Data Reduction and Statistical Analysis

Event markers were created at IC and 200 ms prior to and following IC. Cutting velocity was recorded by observing the pelvis segment peak and average angular velocity. Independent samples *t*-tests were performed to check for differences between the CAI group and the healthy control group.

Kinematic data were exported for peak angular displacement, angular velocity and angular acceleration for the period 200 ms prior to IC to IC (pre-contact), at IC and for the period from IC to 200 ms post contact (impact phase). As with previous studies data were exported as metrics for the forefoot-hindfoot angle, forefoot-tibia angle, hindfoot-tibia angle, hip, knee and trunk angles in all planes of motion. So not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment (De Ridder et al., 2013; Wright et al., 2011). Statistical analysis was performed in SPSS (IBM Corp. Released 2016. IBM SPSS Statistics for Windows, Version 24.0. Armonk, NY: IBM Corp). The matched control limb was compared to the CAI group's affected limb using an independent samples *t*-test. The unaffected and the affected limb of the CAI group were then compared using a paired samples *t*-test ($\alpha = 0.05$). A matched control limb was also compared to the CAI group's unaffected limb using an independent samples *t*-test ($\alpha = 0.05$). Data is presented as group means and standard deviations and the symbol Δ denotes differences between group means.

8.3.3 Results

Independent samples *t*-tests for full kinematic curve analysis revealed no significant differences ($p > 0.05$) between groups for age, height, mass, or shoe size. Independent samples *t*-test also revealed no significant differences between the control group and CAI group groups for average (CAI = 1.30 ± 0.17 m.s^{-1}, control = 1.31 ± 0.12 m.s^{-1}) and maximum (CAI = 1.63 ± 0.21 m.s^{-1}, control = 1.65 ± 0.17 m.s^{-1}) cutting velocity.

CAI Group Affected Limb Versus Matched Control

When analysing the pre-contact phase, a decreased FFTBA internal rotation angular displacement was observed in the affected limb when compared to the matched control (Δ = -4.03°, $p = 0.028$). A decreased FFTBA eversion angular velocity (Δ = 30.09°/s, p =

0.028) and an increased peak positive hip frontal plane angular acceleration (Δ = 631.97˚/s^2, p = 0.046) were also observed in the affected limb when compared to the matched control (Table 8.1). No other significant differences were observed between the affected limb and the matched control pre-contact (Appendix L, Table L.1, Table L.4, Table L.7).

At IC, no significant differences were observed between groups for angular displacement (Appendix L, Table L.2). A decreased FFHFA dorsiflexion angular velocity (Δ = 43.46˚/s, p = 0.046) was observed in the affected limb when compared to the matched control. When investigating angular acceleration an increased FFHFA frontal plane peak positive (Δ = -652.03˚/s^2, p = 0.009), a decreased FFTBA transverse plane peak negative (Δ = 1356.20˚/s^2, p = 0.019) and an increased HFTBA transverse plane peak positive (Δ = 1106.38˚/s^2, p = 0.036) angular acceleration were observed in the affected limb when compared to the matched control (Table 8.1). No other significant differences were observed between the affected limb and the matched control at IC (Appendix L, Table L.5, Table L.8)

During the impact phase, no significant differences were observed between the affected limb of the CAI group, when compared to the matched control for angular displacement (Appendix L, Table L.3) or angular acceleration (Appendix L, Table L.6), however significantly increased FFTBA plantarflexion (Δ = 53.21˚/s, p = 0.041), FFTBA internal rotation (Δ = 31.22˚/s, p = 0.005), HFTBA internal rotation (Δ =32.81˚/s, p = 0.007) and hip external rotation (Δ =23.28˚/s^2, p = 0.049) angular velocity was observed (Table 8.1). No other significant differences were observed in angular velocity during the impact phase between the affected limb and the matched control (Appendix L, Table L.9).

Table 8.1 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the affected limb of the CAI group to a matched control.

AFFECTED LIMB COMPARED TO MATCHED CONTROL			
	200 MS PRE TO INITIAL CONTACT	**INITIAL CONTACT**	**INITIAL CONTACT TO 200 MS POST CONTACT**
ANGULAR DISPLACEMENT	- ↓ FFTBA internal rotation angular displacement (affected: 8.18 ± 5.17, control: 12.21 ± 5.40, p = 0.028)	- No significant differences observed	- No significant differences observed
ANGULAR VELOCITY	- ↓ FFTBA eversion angular velocity (affected: -39.88 ± 33.29, control: -69.97 ± 43.58, p = 0.026)	- ↓ FFHFA dorsiflexion angular velocity (affected: 34.95 ± 67.26, control: 78.41 ± 58.11, p = 0.046)	- ↑ FFTBA plantarflexion angular velocity (affected: -38.18 ± 91.15, control: 15.03 ± 54.32, p = 0.041) - ↑ FFTBA internal rotation angular velocity (affected: 77.00 ± 35.69, control: 45.78 ± 25.60, p = 0.005) - ↑ HFTBA internal rotation angular velocity (affected: 95.53 ± 41.06, control: 62.72 ± 23.49, p = 0.007) - ↑ Hip external rotation angular velocity (affected: -80.91 ± 37.02, control: -57.63 ± 31.19, p = 0.049)
ANGULAR ACCELERATION	- ↑ peak positive hip frontal plane angular acceleration (affected: 2261.46 ± 1093.78, control: 1629.49 ± 689.84, p = 0.046)	- ↑ FFHFA frontal plane angular acceleration (affected: 309.97 ± 960.85, control: -652.03 ± 1125.86, p = 0.009) - ↓ FFTBA transverse plane angular acceleration (affected: -709.89 ± 1785.91, control: -2066.09 ± 1500.00, p = 0.019) - ↑ HFTBA transverse plane angular acceleration (affected: 813.03 ± 1556.73, control: -293.35 ± 1484.53, p = 0.036)	- No significant differences observed

CAI Group Unaffected Limb Versus Matched Control

When comparing the unaffected limb to the healthy matched control during the pre-contact phase, no significant differences were observed in angular displacement (Appendix L, Table L.1). A significantly increased hip flexion angular velocity (Δ = 51.47˚/s, p = 0.035), increased hip abduction angular velocity (Δ = 36.00˚/s, p = 0.011) and increased trunk external rotation angular velocity (Δ = 27.22˚/s, p = 0.008) was observed in the unaffected limb when compared to the matched control. When further comparing hip kinematics between groups, a significantly increased peak positive (Δ = 484.15˚/s^2, p = 0.022) and negative frontal plane motion (Δ = 586.71˚/s^2, p = 0.027) angular acceleration were observed. Alongside this a peak positive trunk sagittal (Δ = 488.16˚/s^2, p = 0.001) and transverse plane (Δ = 442.89˚/s^2, p = 0.024) and a peak negative trunk transverse plane (Δ = -311.69˚/s^2, p = 0.015) angular accelerations were also observed (Table 8.2). No other significant differences were observed in angular velocity and angular acceleration between the unaffected limb and the matched control pre-contact (Appendix L, Table L.4, Table L.7).

At IC, a significantly decreased FFTBA inversion angular displacement (Δ = 24.13˚, p = 0.041) was observed in the unaffected limb when compared to the healthy matched control. When comparing angular velocities, a significantly increased FFHFA inversion (Δ = 35.32˚/s^2, p = 0.022), increased FFTBA inversion (Δ = 35.32˚/s, p = 0.022), increased hip extension (Δ = 30.99˚/s, p = 0.029), increased knee extension (Δ = 110.05˚/s, p = 0.003) and a decreased FFHFA abduction (Δ = 22.28˚/s, p = 0.040) angular velocity were observed in the unaffected limb. A significantly decreased FFTBA transverse plane peak negative (Δ = 1008.41˚/s^2, p = 0.024) and a significantly increased HFTBA transverse plane peak positive (Δ = 794.69˚/s^2, p = 0.027) angular acceleration were also observed in the unaffected limb at IC (Table 8.2). No other significant differences were observed between the unaffected limb and the matched control at IC (Appendix L, Table L.2, Table L.5, Table L.8)

Post IC, no significant differences were observed between groups for angular displacement (Table L.3). A significantly increased FFHFA adduction (Δ = 20.95°/s, p = 0.011) and a significantly increased HFTBA plantarflexion (Δ = 31.48°/s, p = 0.044) angular velocity were observed in the unaffected limb. A significantly decreased peak negative FFTBA transverse plane (Δ = 688.70°/s^2, p = 0.038) angular acceleration and significantly increased peak positive hip sagittal plane (Δ = 783.23°/s^2, p = 0.039), peak positive hip frontal plane (Δ = 698.44°/s^2, p = 0.018), peak positive trunk transverse plane (Δ = 350.65°/s^2, p = 0.043) and peak negative knee sagittal plane (Δ = 1669.60°/s^2, p = 0.033) angular acceleration were observed in the unaffected limb (Table 8.2). No other significant differences were observed during the impact phase between the affected limb and the matched control (Appendix L, Table L.6, Table L.9).

Table 8.2 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the unaffected limb of the CAI group to a matched control.

UNAFFECTED LIMB COMPARED TO MATCHED CONTROL			
	200 MS PRE TO INITIAL CONTACT	**INITIAL CONTACT**	**INITIAL CONTACT TO 200 MS POST CONTACT**
ANGULAR DISPLACEMENT	- No significant differences observed	- ↓ FFTBA inversion displacement (unaffected: -9.98 ± 6.12, control: 14.15 ± 5.64, *p* = 0.041)	- No significant differences observed
ANGULAR VELOCITY	- ↑ Hip flexion angular velocity (unaffected: 89.07 ± 85.36, control: 37.60 ± 50.52, *p* = 0.035) - ↑ Hip Abduction angular velocity (unaffected: -103.79 ± 44.65, control: -67.79 ± 35.46, *p* = 0.011) - ↑ Trunk external rotation angular velocity (unaffected: -48.57 ± 28.48, control: -27.22 ± 15.01, *p* = 0.008)	- ↑ FFHFA inversion angular velocity (unaffected: 70.72 ± 35.65, control: 35.40 ± 50.95, *p* = 0.022) - ↓ FFHFA abduction angular velocity (unaffected: -1.25 ± 29.21, control: -23.53 ± 33.38, *p* = 0.040) - ↑ FFTBA inversion angular velocity (unaffected: 119.35 ± 47.29, control: 68.44 ± 76.57, *p* = 0.022) - ↑ Hip extension angular velocity (unaffected: -68.81 ± 42.72, control: -37.82 ± 38.49, *p* = 0.029) - ↑ Knee extension angular velocity (unaffected: 22.56 ± 116.56, control: -87.49 ± 91.62, *p* =0.003)	- ↑ FFHFA adduction angular velocity (unaffected: 45.70 ± 23.47, control: 24.75 ± 23.06, *p* = 0.011) - ↑ HFTBA plantarflexion angular velocity (unaffected: -22.63 ± 52.23, control: 8.85 ± 38.52, *p* = 0.044)
ANGULAR ACCELERATION	- ↑ peak positive hip frontal plane angular acceleration (unaffected: 2223.65 ± 718.49, control: 1739.50 ± 450.34, *p* = 0.022) - ↑ peak negative hip frontal plane angular acceleration (unaffected: -1894.20 ± 823.34, control: -1307.49 ± 692.13, *p* = 0.027) - ↑ peak positive trunk sagittal plane angular acceleration (unaffected: 1121.28 ± 523.68, control: 633.12 ± 200.54, *p* = 0.001) - ↑ peak positive trunk transverse plane angular acceleration (unaffected: 1286.86 ± 732.24, control: 843.97 ± 245.23, *p* = 0.024) - ↑ peak negative trunk transverse plane angular acceleration (unaffected: -639.99 ± 475.55, control: -328.30 ± 146.68, *p* = 0.015)	- ↓ FFTBA transverse plane angular acceleration (unaffected: -1182.58 ± 911.58, control: -2190.99 ± 1533.81, *p* = 0.024) - ↑ HFTBA transverse plane angular acceleration (unaffected: 358.10 ± 900.12, control: -436.59 ± 1147.63, *p* = 0.027)	- ↓ peak negative FFTBA transverse plane angular acceleration (unaffected: -2673.42 ± 658.90, control: -3362.12 ± 1182.58, *p* =0.038) - ↑ peak positive hip sagittal plane angular acceleration (unaffected: 3084.80 ± 1336.71, control: 2301.57 ± 754.01, *p* = 0.039) - ↑ peak positive hip frontal plane angular acceleration (unaffected: 2703.79 ± 865.74, control: 2005.35 ± 817.61, *p* = 0.018) - ↑ peak negative knee sagittal plane angular acceleration (unaffected: -7106.97 ± 2771.97, control: -5437.37 ± 1475.37, *p* = 0.033) - ↑ peak positive trunk transverse plane angular acceleration (unaffected:1186.02 ± 636.556, control: 835.37 ± 281.90, *p* = 0.043)

CAI Affected Limb Versus CAI Unaffected Limb

The final comparison was between the affected limb when compared to the unaffected limb of the CAI group. No significant differences were observed in angular displacement at any time point analysed in this study (Appendix L, Table L.1, Table L.2,Table L.3).

At pre-contact, several significantly decreased angular velocities were observed in the affected limb when compared to the unaffected limb. Significantly decreased FFHFA plantarflexion (Δ = 25.61°/s, p = 0.046), FFHFA eversion (Δ = 12.32°/s, p = 0.041), FFTBA plantarflexion (Δ = 39.40°/s, p = 0.031), FFTBA eversion (Δ = 19.03°/s, p = 0.033), HFTBA external rotation (Δ = 23.97°/s, p = 0.032) and trunk external rotation (Δ = 14.64°/s, p = 0.015) angular velocities were observed. When observing angular accelerations, a significantly decreased peak positive FFTBA sagittal plane (Δ = 1752.11°/s^2, p = 0.021), a significantly decreased peak negative FFTBA sagittal plane (Δ = 660.05°/s^2, p = 0.046), significantly decreased peak positive knee sagittal plane (Δ = 554.05°/s^2, p = 0.042) and a significantly decreased peak positive trunk transverse plane (Δ = 380.44°/s^2, p = 0.040) angular acceleration were observed in the affected limb (Table 8.3). No other significant differences were observed between groups during the pre-contact phase (Appendix L, Table L.4, Table L.7).

At IC, a significantly increased knee flexion angular velocity (Δ = 53.45°/s, p = 0.025) and a significantly decreased FFHFA sagittal plane peak positive angular acceleration (Δ = 865.74°/s^2, p = 0.047) were observed in the affected limb when compared to the unaffected limb (Table 8.3). No other significant differences were observed between the affected and unaffected limb at IC (Appendix L, Table L.5, Table L.8).

Following IC, a significantly decreased FFHFA dorsiflexion (Δ = 30.52°/s, p = 0.041), decreased hip flexion (Δ = 23.30°/s, p = 0.028) and decreased knee extension (Δ =

35.48°/s, p = 0.033) angular velocities were observed in the affected limb when compared to the unaffected limb. A significantly decrease peak negative FFHFA sagittal plane (Δ = 419.41°/s^2, p = 0.046), decreased peak negative FFTBA sagittal plane (Δ = 588.43°/s^2, p = 0.046), increased peak negative FFTBA transverse plane (Δ = 707.60°/s^2, p = 0.037) and decreased peak negative hip sagittal plane (Δ = 468.11°/s^2, p = 0.030) angular accelerations were also observed in the affected limb when compared to the unaffected (Table 8.3). No other significant differences were observed between the affected and unaffected limb during this time point (Appendix L, Table L.6, Table L.9).

Table 8.3 Significant differences observed in angular displacement (˚), angular velocity (˚/s) and angular acceleration (˚/s^2) when comparing the affected limb to the unaffected limb of the CAI group.

AFFECTED LIMB COMPARED TO UNAFFECTED LIMB			
	200 MS PRE TO INITIAL CONTACT	**INITIAL CONTACT**	**INITIAL CONTACT TO 200 MS POST CONTACT**
ANGULAR DISPLACEMENT	- No significant differences observed	- No significant differences observed	- No significant differences observed
ANGULAR VELOCITY	- ↓ FFHFA plantarflexion angular velocity (unaffected: -129.38 ± 41.33, affected: -103.77 ± 42.59, *p* = 0.046) - ↓ FFHFA eversion angular velocity (unaffected: -42.31 ± 20.87, affected: -29.99 ± 22.98, *p* = 0.041) - ↓ FFTBA plantarflexion angular velocity (unaffected: -252.57 ± 65.89, affected: -213.17 ± 74.15, *p* = 0.031) - ↓ FFTBA eversion angular velocity (unaffected: -58.91 ± 24.16, affected: -39.88 ± 33.29, *p* = 0.033) - ↓ HFTBA external rotation angular velocity (unaffected: -87.76 ± 54.46, affected: -63.79 ± 43.39, *p* = 0.032) - ↓ Trunk external rotation angular velocity (unaffected: -48.57 ± 28.48, affected: -33.93 ± 26.22, *p* = 0.015)	- ↑ Knee flexion angular velocity (unaffected: 22.56 ± 116.56, affected: -30.89 ± 129.63, *p* = 0.025)	- ↓ FFHFA dorsiflexion angular velocity (unaffected: 189.64 ± 54.76, affected: 159.12 ± 53.59, *p* = 0.041) - ↓ Hip flexion angular velocity (unaffected: -132.51 ± 70.79, affected: -109.21 ± 73.96, *p* = 0.028) - ↓ Knee extension angular velocity (unaffected: 98.17 ± 124.63 ˚/s, affected: 62.69 ± 114.17, *p* = 0.033)
ANGULAR ACCELERATION	- ↓ peak positive FFTBA sagittal plane angular acceleration (unaffected: 9195.40 ± 2184.69, affected: 7443.29 ± 3131.21, *p* = 0.021) - ↓ peak negative FFTBA sagittal plane angular acceleration (unaffected: -4400.89 ± 2531.33, affected: -3740.84 ± 2408.84, *p* = 0.046) - ↓ peak positive knee sagittal plane angular acceleration (unaffected: 4607.73 ± 2171.51, affected: 4053.68 ± 1915.97, *p* = 0.042) - ↓ peak positive trunk transverse plane angular acceleration (unaffected: 1286.86 ± 732.24, affected: 906.42 ± 563.79, *p* = 0.040)	- ↓ FFHFA sagittal plane angular acceleration (unaffected: 4103.52 ± 1841.49, affected: 3237.78 ± 1462.76, *p* = 0.047)	- ↓ peak negative FFHFA sagittal plane angular acceleration (unaffected: -2918.65 ± 749.43, affected: -2499.24 ± 980.90, *p* = 0.046) - ↓ peak negative FFTBA sagittal plane angular acceleration (unaffected: -4114.41 ± 1298.32, affected: -3525.98 ± 1869.56, *p* = 0.046) - ↑ peak negative FFTBA transverse plane angular acceleration (unaffected: -2673.42 ± 658.90, affected: -3381.02 ± 1592.25, *p* = 0.037) - ↓ peak negative hip sagittal plane angular acceleration (unaffected: -3064.75 ± 1015.85, affected: -2596.64 ± 877.77, *p* = 0.030)

8.3.4 Discussion

This study aimed to identify differences in discrete kinematic variables during cutting manoeuvres between CAI participants' unaffected and affected ankles and to matched control limbs. As significant differences were observed in kinematics null hypothesis $H0_{11}$ can be rejected.

CAI group affected limb versus matched control

Prior to IC, a decreased FFTBA internal rotation angular displacement was observed as with Chapter 6.3 and 7.3. As previously discussed in this thesis, it is proposed that this may be either a protective mechanism as internal rotation is now suggested to be involved in lateral ankle sprain mechanisms (Fong et al., 2009a; Fong et al., 2012; Konradsen & Voigt, 2002). It is, however, also postulated that this small reduction in internal rotation may also place individuals at increased risk of injury due to the changes in joint arthrokinematics that occur as a result of this position. It is suggested that a slight increase in internal rotation may create a more closed-packed position thereby restricting the degree of subtalar inversion available. Kirby (2001) suggested that the positioning of the subtalar joint axis dictates the magnitude and direction of pronation and supination moments. When this axis is more medially deviated it is suggested increased pronation and decreased supination moments are likely. Prospective research is needed into this variable to determine whether this is something that may predispose individuals to the initial ankle sprain or whether this modification may be because of the sprain. Further research should investigate whether internal rotation is blocked, potentially due to differences in bone shape or positioning that have been documented in those with ankle instability (Tümer et al., 2019) or whether this is an active modification.

Also, during the initial pre-contact phase, a decreased FFTBA eversion angular velocity was observed. Needle et al. (2014) suggested that prior to ground contact, a feedforward preparatory strategy should be adopted by the limb whereby muscle tension is produced in preparation for the ground contact, this increases joint stiffness and also increases alpha-gamma co-activation enabling quicker identification of length change. It is speculated that this reduction in eversion angular velocity may be as a result of damage to capsuloligamentous mechanoreceptors following injury, therefore, reducing the feedforward strategy prior to IC (Needle et al., 2014). Decreased peroneal activation and reaction time have previously been observed in affected ankles (Delahunt et al., 2006a; Mitchell et al., 2008), which may also explain the reduction in eversion angular velocity. Increased peak positive hip frontal plane angular acceleration were also observed in the affected limb when compared to the matched control further research may be necessary to identify the exact cause of this finding.

Interestingly, at IC, no significant differences were observed in angular displacement between the affected and the matched control limb. This highlights the need to investigate the speed at which movement occurs when comparing between injured and healthy groups. A decreased FFHFA dorsiflexion angular velocity was observed in the affected limb when compared to the healthy matched control. This may indicate a less effective protective strategy implemented to move out of the plantarflexed position previously associated with lateral ankle sprains (Fong et al., 2009b). This finding combined with landing on an unstable surface or on an opponent's foot (a common mechanism for lateral ankle sprains) may present an increased injury risk. An increased FFHFA frontal plane peak positive, a decreased FFTBA transverse plane peak negative and an increased HFTBA transverse plane peak positive angular acceleration were observed in the affected limb when compared to the matched control, potentially highlighting increased correction of movement on IC.

For the impact phase, no significant differences were observed in angular displacement or angular acceleration. A significantly increased FFTBA plantarflexion, FFTBA internal rotation, HFTBA internal rotation and hip external rotation angular velocity were observed in the affected limb. This significant increase in the speed of movement modification is suggestive of a poor feedforward mechanism in preparation for ground contact as observed during previous research of dynamic movements (Caulfield & Garrett, 2004; Lin et al., 2011; K. A. Webster et al., 2016). Alternatively, it may signpost towards a reduction in neuromuscular control at the joint (Delahunt et al., 2006a; Gutierrez et al., 2009). Previous research has observed an increase in hip flexion, hip abduction and knee flexion during the stance phase of a 45-degree cut (Koshino et al., 2014). Differences in findings may be due to the difference in cutting angle changing the demands on the body. Differences in foot kinematics may also be due to the more detailed foot model implemented within this study when compared to the work of Koshino et al (2016) and Son et al. (2017)

CAI group unaffected limb versus matched control

When comparing the unaffected limb of the CAI group to the healthy matched control, no significant differences were observed in angular displacement prior to IC. When analysing angular velocity, a significantly increased hip flexion, hip abduction and trunk external rotation angular velocity, were observed in the unaffected when compared to the matched control. When comparing angular acceleration between limbs, a significantly increased peak positive and negative hip frontal plane, increased peak positive trunk sagittal and transverse plane and increased peak negative trunk transverse plane angular accelerations were observed. These combined findings suggest a modified feedforward approach (Caulfield & Garrett, 2004; Lin et al., 2011; K. A. Webster et al., 2016) prior to IC on the unaffected limb – this could be speculated as an attempt to increase the stability of the unaffected limb but may also be due to the affected limb being the last one in contact with the ground. No research at present compares the unaffected limb to a healthy matched control limb during a cut. The presence of differences between groups raises into question the validity of research that uses this limb as the “control” limb. It also highlights the need for bilateral rehabilitation and biomechanical screening prior to return to play following initial lateral ankle sprains.

At IC, a significantly decreased FFTBA inversion angular displacement was observed in the unaffected limb compared to the healthy matched control. When comparing angular velocities, a significantly increased FFHFA inversion angular velocity was observed. When comparing the values of the two matched control limbs there seemed to be a large difference between these values. It is suggested therefore that this finding may be due to a reduction in the matched control score and therefore should be interpreted with a degree of caution. Increased FFTBA inversion, increased hip extension, increased knee extension and a decreased FFHFA abduction angular velocity was observed in the unaffected limb. A significantly decreased FFTBA peak negative transverse plane and a significantly increased HFTBA peak positive transverse plane angular acceleration were also observed in the unaffected limb at IC. It has been suggested that pain can often be associated with redistribution of activity between muscles, increased stiffness or modification of movement and efforts to ‘splint’ a joint to protect from further pain or injury (Hodges, 2011). This theory goes further to suggest that most adaptations occur on a subconscious level however some may involve the nervous system (withdrawal reflex) or higher-level processing and planning and even voluntary adjustments. It is possible that these adaptations observed may be preparatory for the next step onto the affected limb. As with the findings prior to IC, it is suggested that bilateral rehabilitation and biomechanical screening prior to returning to play following initial lateral ankle sprains is required.

Following IC, no significant differences were observed between groups for angular displacement. A significantly increased FFHFA adduction and a significantly increased HFTBA plantarflexion angular velocity were observed; this fast modification of foot positioning may explain the reasoning for high incidence of bilateral instability (Tanen et al., 2014). A significantly decreased peak negative FFTBA transverse plane angular acceleration and significantly increased peak positive hip sagittal, peak positive hip frontal plane, peak positive trunk transverse plane and peak negative knee sagittal plane angular acceleration were observed in the unaffected limb. The large number of differences observed between the unaffected limb and the healthy matched control highlight the importance of addressing both limbs during rehabilitation. They also highlight the possible reasons behind the high incidence of bilateral instability (Tanen et al., 2014) as different movement strategies are adopted by the unaffected limb. This may suggest that central issues (Doherty et al., 2016b; Hass, Bishop, Doidge, & Wikstrom, 2010) may be present in individuals with CAI potentially predisposing to bilateral lateral ankle sprains. It is unclear as to the reasoning behind some of these strategies, however it is suggested that it may be an attempt to alter the position of the body's centre of gravity (Delahunt et al., 2007), an attempt to increase joint stiffness in order to protect the affected limb or due to a lack of control.

CAI affected limb versus CAI unaffected limb

When comparing the affected limb to the unaffected limb of the ankle instability group, no significant differences were observed in angular displacement at any time point analysed in this study. During the pre-contact phase of the cutting manoeuvre, significantly decreased FFHFA plantarflexion, FFHFA eversion, FFTBA plantarflexion and FFTBA eversion angular velocity was observed in the affected limb compared to the unaffected limb. A decrease in FFHFA and FFTBA eversion angular velocity may signify an inability to identify an "at risk" foot position in feedforward preparation of foot contact (Needle et al., 2014). The reduction of plantarflexion angular velocity may be an attempt by the joint to increase talocrural joint stiffness in preparation for the ground contact potentially as a protective mechanism (Hodges, 2011). Significant differences were also observed in HFTBA external rotation and trunk external rotation angular velocities where decreased angular velocities were observed in the affected limb. On further inspection, however, these differences seem to be due to increased velocity in the unaffected limb when compared to the affected limb. This would appear to be a protective mechanism to move away for the affected limb, though further research may be necessary into this area, investigating the displacement of the centre of mass as this may provide further insight into the cause of this finding. When observing angular accelerations, a significantly decreased peak positive FFTBA sagittal plane, a significantly decreased peak negative FFTBA sagittal plane, significantly decreased peak positive knee sagittal plane and a significantly decreased peak positive trunk transverse plane angular acceleration were also observed. It is unclear as to the reasons for these differences.

At IC, a significantly increased knee flexion angular velocity and a significantly decreased FFHFA sagittal plane peak positive angular acceleration were observed in the affected limb compared to the unaffected limb. This may suggest an attempt to lower the position of the body's centre of gravity (Gribble & Robinson, 2009, 2010) using proximal adaptations as a method to improve joint stability.

The final phase of this movement (post-contact), observed a significantly decreased FFHFA dorsiflexion, decreased hip flexion and decreased knee extension angular velocity in the affected limb when compared to the unaffected limb. This may be due to the already lowered position of the body's centre of gravity though further research would be beneficial to determine whether this is the case and use of quadriceps muscle activation patterns may also be beneficial as it is suggested that these may be increased. A significantly decreased peak negative FFHFA sagittal plane, decreased peak negative FFTBA sagittal plane, increased peak negative FFTBA transverse plane and decreased peak negative hip sagittal plane angular accelerations were also observed in the affected limb when compared to the unaffected. It is unclear as to the reasons for these differences.

Clinical implications

When comparing FFTBA internal rotation between the affected limb and the matched control, as with the walk and landing movements, a decrease in internal rotation was observed (Δ = 4.03°). The observation that this is the case across all three movements covered in this thesis each at the same time point suggests that this is an area for future investigation and potentially an area to target future interventions. At the same time point a decreased FFTBA eversion angular velocity was observed (Δ =30.09°/s) previous literature has reported links between a decreased eversion velocity and injury history (Kuhman, Paquette, Peel, & Melcher, 2016) suggesting this difference may be of clinical significance.

When comparing the unaffected to the affected limb no significant differences were observed in angular displacements. Instead all 19 differences were observed in angular velocities and accelerations suggesting the rate that an angular displacement is covered may be of more clinical interest.

Several significant differences were highlighted between the affected limb and the unaffected limb of the CAI group and when comparing to the healthy matched control. This stresses the need for bilateral rehabilitation when treating individuals both following an initial acute lateral ankle sprain but also when working with athletes with CAI. It also highlights the flaws of a number of research papers where the unaffected limb is used for comparisons as a "control" limb. The findings in the current study also call for the implementation of full-body exercise interventions incorporating the entire kinetic chain when rehabilitating athletes. Full body kinematic bilateral analysis is also essential in the rehabilitation of individuals during common injury provoking manoeuvres.

Limitations

As with previous studies, the purpose of this research was to investigate the kinematics of movement between individuals with ankle instability and healthy controls. Force data were not included within this study and as such there are limitations regarding the conclusions that the authors are able to make with regard to change in centre of mass and joint moments.

Future research

Future research should incorporate full body kinematic and kinetic analysis with the use of the multi-segmental foot model to enable a more in-depth analysis to be undertaken. It also remains unclear as to whether the differences highlighted in this study were present prior to the initial lateral ankle sprain that was obtained or as a compensatory strategy following the initial ankle sprain. As such the author would suggest the need for a prospective study of athletes prior to ankle sprains, tracking movement mechanics and cross-relating these findings to the incidence of injury. Future research should also continue to address the angular velocities of movements as more clinically significant findings may be obtained.

8.3.5 Conclusion

Several significant differences have been identified between individuals with CAI and healthy controls. Differences have also been identified in the unaffected limb and higher up the kinetic chain suggesting that recurrent ankle sprains not only affect the ankle joint but may also result in central neuromuscular issues and problems with the unaffected limb. This information is crucial when creating preventative and rehabilitative strategies for ankle sprains.

8.3.6 Development of Research

As identified in the first 2 studies (Chapter 6.3 and 7.3) a decreased FFTBA internal rotation angular displacement was observed prior to IC. This may provide a key consistent variable for preventative and rehabilitation strategies.

This study has also highlighted several variables have been highlighted as significantly different when comparing the affected limb of the CAI group to a healthy matched control during a cutting manoeuvre. The next section of this 3 part study will perform a regression analysis to identify whether these variables are correlated to score on the IdFAI questionnaire and whether these variables may be used to predict scores on this questionnaire. This may aid in further validating the questionnaire, developing objective markers to use together with the questionnaire to identify ankle instability and also highlighting key areas that may need additional rehabilitation due to higher scores on the IdFAI questionnaire.

8.4 Study 3 – Part 3 – Can Significantly Different Kinematic Variables Observed During a Cutting Manoeuvre Between Individuals with and Without Ankle Instability Predict Identification of Functional Ankle Instability Questionnaire Score?

8.4.1 Introduction

As highlighted in Chapter 6.4 no clear globally accepted measure exists for the classification of ankle instability (Simon et al., 2014). As such this is normally of a self-reported nature with the use of questionnaires. The International Ankle Consortium published selection guidelines and cut-off score values for identification of ankle instability (Gribble et al., 2013). Each of these questionnaires places individuals on a scale, however, the cut-off value is often used to determine a 'healthy' individual and one with instability. The preceding chapter identified several differences in angular displacements, angular velocities and angular accelerations during cutting manoeuvres between individuals with ankle instability and healthy controls. This study will identify whether the previously identified variables are correlated to the score individuals get on the IdFAI questionnaire and subsequently whether these variables may be used as a more objective measure to identify ankle instability.

This study will address the following hypothesis previously presented in Chapter 2.19:

- **H_{12}** - Significant differences observed during cutting will be able to predict IdFAI questionnaire score

8.4.2 Methods

Variables where significant differences were observed between the affected limb of the CAI group and the healthy matched control limb during the cutting manoeuvre (study 3, part 2 (8.3.3)), were used for analysis. These are outlined in Table 8.4.

Table 8.4 Variables with significant differences in study 3 part 2 used within the regression analysis.

	100 ms pre- heel strike	**Heel strike**	**Heel strike – 200 ms post heel strike**
Angular displacement	- ↓ FFTBA internal rotation	-	-
Angular velocity	- ↓ FFTBA eversion	- ↓ FFHFA dorsiflexion	- ↑ FFTBA plantarflexion - ↑ FFTBA internal rotation - ↑ HFTBA internal rotation - ↑ Hip external rotation
Angular acceleration	- ↑ peak positive hip frontal plane	- ↑ FFHFA frontal plane - ↓ FFTBA transverse plane - ↑ HFTBA transverse plane	-

Statistical Analysis

Data were assessed for normality using the Shapiro-Wilks test and Pearson's correlation analyses were performed to identify relationships between IdFAI questionnaire score and kinematic variables. All predictors that showed a moderate (r = 0.3 upwards; Cohen & Cohen, 1977) and significant correlation ($p < .05$) were kept for further analysis, while all other variables were removed from subsequent analysis. Linear or multiple stepwise regressions were then used to identify which kinematic variables during cutting best predicted IdFAI score. Data were inspected to ensure it met all assumptions of a regression before proceeding as outlined in 6.4.2.

8.4.3 Results

Significant correlations can be observed in Table 8.5. Eleven variables were included in the initial analyses. Of these variables, 7 were inputted into the stepwise regression analysis.

The stepwise regression analysis showed FFTBA transverse plane transverse acceleration at IC to be the best independent predictor of IdFAI score (R = 0.490, R^2 = 0.240, F = 10.742, p = 0.002). Combined FFTBA transverse plane transverse acceleration at IC and FFTBA frontal eversion velocity prior to IC improved the prediction (R = 0.600, R^2 = 0.360, F = 6.191, p = 0.003). The final model produced included FFHFA frontal acceleration at IC (R = 0.666, R^2 = 0.443, F = 4.756, p =0.037). Prediction equations are outlined in Table 8.6. Beta and standard error values are outlined in Table 8.7.

Table 8.5 Pearson's correlation outputs for kinematic variables compared to IdFAI score

		FFTBA transverse internal rotation displacement 200-IC	HIP frontal max acceleration 200-IC	FFTBA frontal Eversion velocity 200-IC	FFHFA frontal acceleration ICVALUE	FFTBA transverse acceleration ICVALUE	HFTBA transverse acceleration ICVALUE	FFHFA sagittal dorsiflexion velocity ICVALUE	FFTBA sagittal plantarflexion velocity IC-200	FFTBA transverse internal rotation velocity IC-200	HFTBA IR velocity IC-200	Hip transverse IR velocity IC-200
IdFAI SCORE	Pearson Correlation	-.382*	0.302	.393*	.453**	.490**	.471**	-.346*	-0.112	0.267	.351*	-0.121
	Sig. (2-tailed)	0.021	0.073	0.018	0.006	0.002	0.004	0.039	0.515	0.115	0.036	0.483

Table 8.6 Prediction equations for IdFAI score

Model	Equation
Model 1	IdFAI Score = (0.152 x FFTBA transverse acceleration ICVALUE) + 15.480 (±8.435)
Model 2	IdFAI Score = (0.141 x FFTBA transverse acceleration ICVALUE) + (0.081 x FFTBA frontal Eversion velocity 200-IC) + 19.654 (± 7.857)
Model 3	IdFAI Score = (0.115 x FFTBA transverse acceleration ICVALUE) + (0.073 x FFTBA frontal Eversion velocity 200-IC) + (0.145 x FFHFA frontal acceleration ICVALUE) + 19.029 (±7.445)

Table 8.7 Unstandardized and standardized Beta values for each of the 9 regression models

Dependent variable	Variable	B	SE	ß	st error of estimate
IdFAI Score (Model 1)	Constant	15.480	1.798		
	FFTBA transverse acceleration ICVALUE	0.152	0.046	0.490	8.435
IdFAI Score (Model 2)	Constant	19.654	2.370		
	FFTBA transverse acceleration ICVALUE	0.141	0.043	0.456	
	FFTBA frontal Eversion velocity 200-IC	0.081	0.032	0.348	7.857
IdFAI Score (Model 3)	Constant	19.029	2.264		
	FFTBA transverse acceleration ICVALUE	0.115	0.043	0.372	
	FFTBA frontal Eversion velocity 200-IC	0.073	0.031	0.314	
	FFHFA frontal acceleration ICVALUE	0.145	0.066	0.302	7.445

8.4.4 Discussion

As alluded to previously in this thesis the IdFAI questionnaire is often used by clinicians and within research for the identification of ankle instability using a scale approach to either rule in or out ankle instability. Anything below the cut off value is deemed as "healthy" whilst anything above is deemed to be unstable. The authors of this present study suggest that this score may be used as a continuum and as such have investigated whether significantly different variables during cutting may be used to predict the 'severity' of this instability.

Of the 11 variables included in the initial analyses, 7 were significantly correlated to the IdFAI questionnaire score. The regression analysis produced 3 models. The first used FFTBA transverse acceleration at IC and was able to predict 24% of the variance. The correlation between the IdFAI score and FFTBA transverse acceleration at IC showed that as IdFAI score increased so too did the FFTBA transverse acceleration. The lowest scores on the IdFAI questionnaire reported negative values for transverse acceleration whilst higher scores on the IdFAI questionnaire reported positive scores. This may signify a faster positive movement or a slower negative. Further research should investigate this acceleration finding further as it remains unclear as to the reason for this whether this is protective or predisposes individuals to potential injury.

The second model included FFTBA frontal eversion velocity prior to initial IC. This increased the percentage to 36%. The correlation between the IdFAI score and frontal eversion velocity prior to initial IC showed that as IdFAI score increased so too did the FFHFA frontal acceleration. Faster eversion was observed in individuals with the lowest scores on the IdFAI questionnaire this may be suggestive of increased protective strategies prior to IC in individuals with more stable ankles.

The final model also incorporated FFHFA frontal acceleration at IC increasing the percentage to 44.3%. The correlation between the IdFAI score and frontal eversion velocity prior to initial foots FFHFA frontal acceleration at IC showed that as IdFAI score increased so too did the FFHFA frontal acceleration. As with the first model, the lowest scores on the IdFAI questionnaire reported negative values for frontal acceleration whilst higher scores on the IdFAI questionnaire reported positive scores. Again, this finding should be further investigated to determine the cause of this to better understand the implications this has on ankle kinematics. As the variables outlined in these models were able to help predict the IdFAI questionnaire score $H0_{12}$ can be rejected.

Clinical implications

Further research should be conducted to ratify whether the variables outlined in this regression model may be used to increase the specificity and sensitivity of tests to confirm ankle instability. The importance of these variables may be key when working with individuals with high scores on the IdFAI questionnaire.

8.4.5 Conclusion

FFTBA transverse acceleration at IC, FFTBA frontal eversion velocity prior to initial IC and FFHFA frontal acceleration at IC were able to account for 44.3% of the IdFAI score variance. This shows a high prediction ability with just three variables.

Chapter 9.0 Discussion

9.1 Thesis Overview

This discussion will highlight the aims and key findings of the 3 main studies of this research. It will also collate key findings that have been observed across each of the movements studied. Clinical implications will be discussed along with the contribution to the literature, limitations and directions for future research.

9.2 Aims of Each Stage

There were three main aims for each of the movements studied within this thesis:

1) To compare movement patterns of the trunk, hip, knee and multi-segmental foot kinematics and muscle activation between participants with CAI and healthy controls
2) To compare angular displacement, angular velocity and angular acceleration of the trunk, hip, knee and foot between participants with CAI and healthy controls
3) To identify whether it is possible to predict scores on the Identification of Functional Ankle Instability questionnaire from the identified variables of significant differences

Due to the exploratory nature of this thesis, general study hypotheses were proposed. For the walking movement these were that CAI participants would display modified kinematic (H_1) and modified muscle activation patterns (H_2), modified discrete kinematic variables (H_3) and that the significant differences observed would be able to predict the score on the IdFAI questionnaire (H_4). These were echoed in the single leg landing (H_5, H_6, H_7 and H_8) and the cutting manoeuvre (H_9, H_{10}, H_{11} and H_{12}). The null hypotheses (presented in Chapter 2.19) were that no significant differences would be observed and that prediction of the IdFAI questionnaire would not be possible (H0).

9.3 Review of Hypotheses

With reference to the null hypotheses outlined in section 2.19, the $H0_1$, $H0_3$, $H0_5$, $H0_7$, $H0_9$ and $H0_{11}$ can all be rejected, as differences in kinematics were observed between groups for the walk, land and cut manoeuvres. Null hypotheses $H0_2$ and $H0_6$ cannot be rejected as no significant differences were observed in muscle activation patterns during the walk and landing manoeuvres. Comparatively, $H0_{10}$ can be rejected as a significant difference was observed in peroneus longus muscle activation patterns during cutting. Lastly null hypotheses $H0_4$, $H0_8$ and $H0_{12}$ can also be rejected as models produced in the walking manoeuvre were able to predict 49.6%, in the land 35.2% and in the cut 44.3%.

9.4 Key Findings

9.4.1 Statistical Parametric Mapping

Aim - 'To compare movement patterns of the trunk, hip, knee and multi-segmental foot kinematics and muscle activation between participants with CAI and healthy controls'

The above aim (to compare movement patterns) was achieved with the statistical analysis method of statistical parametric mapping. Below is a summary of the significant differences observed between each movement (Table 9.1). A significant difference in FFTBA frontal plane motion was identified, where increased inversion was observed in the walking (mean difference = 3.07°, peak difference = 3.24°) and the cutting movements (mean difference = 3.71°, peak difference = 3.76°) in the affected limb when compared to the unaffected limb. This difference was observed in the stance phase of walking (4-16%) and also in the pre-contact phase of cutting (56-73%). Total range of motion in the frontal plane has previously been reported to be 35 degrees (Brockett & Chapman, 2016) thus differences of approximately 3 degrees in both the walk and cutting movements suggests this variable is of key interest. The difference in movement patterns observed may suggest compensatory strategies adopted by those with CAI. Increased inversion is known to decrease bony restrictions of the foot-ankle complex thus having the potential to increase inversion torque and joint susceptibility to injury (Konradsen, 2002). This may predispose the individual to recurrent sprains or alternatively be a factor that made the individual susceptible to the initial sprain. In walking, this was the only difference in movement and muscle activation patterns observed. Differences in hip kinematics of the unaffected limb (increased hip abduction) when compared to the matched control, were unique to the landing movement. This may be due to the dynamic balance and postural control element necessary in order to stabilise the movement (Durall et al., 2011). The only movement to observe differences between the affected limb and the matched control limb was the cutting manoeuvre. Where a decreased FFTBA internal rotation was observed in the affected limb at 0-11% pre initial contact (mean difference 4.98°). Previous cadaveric studies have observed maximal internal rotation range of 17.62° (Wilkerson et al., 2010), thus a mean difference of 4.98° may have a clinically significant impact on movement adopted. Further research is needed to determine whether this is a modification of movement as a direct result of an initial ankle sprain or a movement pattern already adopted and whether a lack of range of motion was present within the ankle joint itself.

Table 9.1 Summary of SPM findings from walking, single leg landing and cutting manoeuvres

	Affected - Control	Unaffected - Control	Affected - Unaffected
Walking			Significantly increased FFTBA inversion in the affected limb at 4-16% of the stance phase (mean difference = 3.07˚, peak difference = 3.24˚, *p* = 0.039)
Single Leg Landing		Significantly increased hip abduction in the unaffected limb at 30-100% of the impact phase of landing (, mean difference = 4.28˚, peak difference = 4.47˚, *p* = 0.011)	
Cutting	Significantly decreased FFTBA internal rotation in the affected limb at 0-11% of the pre-contact phase (mean difference = 4.98°, peak difference = 5.40°, *p* = 0.048)	Significantly decreased FFTBA inversion in the unaffected limb at 68-90% of the pre-contact phase (mean difference = 5.28°, peak difference = 5.42°, *p* = 0.044).	Significantly increased FFTBA inversion in the affected limb at 56-73% of the pre-contact phase (mean difference = 3.71°, peak difference = 3.76°, *p* = 0.045).
		Significantly increased peroneus longus activation in the unaffected limb at 0-20% of the pre-contact phase (mean difference = 36.04%, peak difference = 3 8.81%, *p* = 0.014)	
		Significantly decreased peroneus longus activation in the unaffected limb from 49-64% of the initial contact to toe off phase (mean difference = 64.50%, peak difference = 71.38%, *p* = 0.001)	

9.4.1.1 Clinical Implications

The fact that most differences were observed in the unaffected limb when compared to the matched control may indicate potential protective strategies. This may be an effort to move the centre of gravity in order to maintain ankle joint stability (Delahunt et al., 2007) and in doing so move away from the affected limb. This does call into question the fact that in rehabilitation environments, movement is commonly compared to the 'good side' and this approach is sometimes adopted as a control within research proposals. Chronic Ankle Instability is often a bilateral issue (Yeung et al., 1994) and therefore this should be considered when devising rehabilitation strategies.

9.4.2 Discrete Variable Analysis

Aim - 'To compare angular displacement, angular velocity and angular acceleration of the trunk, hip, knee and foot between participants with CAI and healthy controls'

The second aim of this thesis (to compare discrete variables) used statistical analysis to determine differences in angular displacement, angular velocity and angular accelerations during walking, landing and cutting. Below is a summary of the significant differences that were observed in each movement (Table 9.2). Of note, is the decrease in FFTBA internal rotation angular displacement, observed in each of the movements (walking = Δ 4.95°, landing = Δ 3.89°, cutting = Δ 4.03°) prior to IC when comparing the affected limb to a healthy matched control. Within this thesis, it was initially proposed that this may be a protective mechanism (Chinn et al., 2013), adopted in order to try to reduce the risk of further lateral ankle sprains, or due to joint restrictions (Vicenzino et al., 2006). However, this may also predispose individuals to ankle sprains due to the arthrokinematics potentially decreasing the restriction on inversion in this position. Kirby (1989, 2001) suggested that the deviation of the subtalar joint axis may hinder rotational equilibrium (sum of the moments acting across that joint axis equal to 0) and therefore manifest in either increased or decreased supination moments. Subtalar joint deformities have previously been linked to ankle joint instability and further the development of ankle joint osteoarthritis (Krähenbuhl, Horn-Lang, Hintermann, & Knupp, 2017). There is currently a paucity of literature to support the finding of decreased FFTBA internal rotation in individuals with ankle instability, therefore this should be investigated further. It is also unclear as to whether the differences observed between groups is present prior to the initial lateral ankle sprain or whether these adaptations occur as compensatory adjustments following the initial incident.

Forefoot-tibia internal rotation angular displacement was the only variable where a significant difference was observed across multiple movements at the same time point. This presents the possible conclusion that due to the differing demands of each of the selected movements, different compensatory strategies or strategies that predisposed the individuals to the initial sprain may exist. As such, it is suggested that with regard to rehabilitation of individuals, a one size fits all approach such as ankle bracing and taping whereby movement is prevented in the same way for all movements completed within a sport may not be the most efficient preventative measure. It may also be necessary to focus on the key movement that a non-contact ankle sprain occurred in. It is suggested that dynamic rehabilitation focusing on retraining with manipulation of the surface may be an effective way to achieve this.

Key differences are observed not just between the affected limb and the matched control, but also between the unaffected limb and a matched control and between the unaffected and affected limb of the CAI group. This suggests that methodology that adopts the contralateral limb as a 'control' is ineffective and that rehabilitation programmes must ensure that due care and attention is paid to the unaffected limb where possible, as protective or compensatory strategies may be adopted.

Summary tables of the walk, land and cut manoeuvres between the affected and matched control (Table 9.2), the unaffected and matched control (Table 9.3) and the unaffected and affected limb of the CAI group (Table 9.4) are presented to enable comparison between movements, highlighting recurrent themes.

Of the 16 differences in angular displacement observed 10 of these were found to be in the forefoot-tibia and 4 of these in the trunk. When comparing the affected limb of the CAI group to the matched control limb 7 differences were observed in angular displacement, 18 in angular velocities and 13 in angular accelerations. When looking at the location of these differences during different movements a total of 9 differences in angular displacements, velocities and accelerations were observed at the foot in the walk and cut movements compared to only 3 in the land. When looking at the rest of the kinetic chain (knee, hip, trunk) 6 differences were observed during the walk, 2 in the cut and 9 during the land. This suggests that more proximal adaptations may be adopted during the landing movement a task concerned with maintenance of the centre of mass within the base of support. This is consistent when comparing between each group, with 9 differences in the trunk, hip and knee compared to 1 in the foot when comparing the unaffected and the matched control and 5 compared to 0 when comparing the affected to unaffected comparison. The number of differences in the foot compared to the rest of the body during the walk are much more equally split (affected – control = 9 foot vs 6 trunk, hip & knee, unaffected – control = 4 foot vs 3 trunk, hip & knee, affected – unaffected = 5 foot vs 4 trunk, hip &knee). The observed deviation in terms of location of modifications in movement pattern suggest that it may be more conducive to design rehabilitation and preventative measures to target multiple movements and a one size fits all approach may be ineffective.

Table 9.2 Affected - Matched control comparisons for angular displacement, angular velocity and angular acceleration between movements

		ANGULAR DISPLACEMENT									ANGULAR VELOCITY									ANGULAR ACCELERATION								
		Walk			Land			Cut			Walk			Land			Cut			Walk			Land			Cut		
		Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post
Forefoot-hindfoot	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓DF	-	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ +VE	-
	Z	-	-	-	-	-	-	-	-	-	-	✓ ↓ABD	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
Forefoot-tibia	X	-	-	-	-	-	-	-	-	-	✓ ↓DF	-	-	-	-	-	-	-	✓ ↑PF	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓EV	-	-	-	✓ ↓ -VE	-	-	-	-	-	-	-
	Z	✓ ↓IR ↓ER	-	✓ ↓IR	✓ ↓IR	-	-	✓ ↓IR	-	-	-	✓ ↓ER	✓ ↓ER	-	-	-	-	-	✓ ↑IR	✓ ↑ +VE	-	-	-	-	-	-	✓ ↓-VE	-
Hindfoot -tibia	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑EV	✓ ↑EV	-	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑IR	-	-	-	-	-	-	-	✓ ↑ +VE	-
Trunk	X	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑FLEX	✓ ↑FLEX	-	-	-	-	-	-	-	✓ ↑ -VE	-	-	-	-	-
	Y	✓ ↓LAT FLEX	-	✓ ↓ LAT FLEX	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓ -VE	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ -VE	-	-	-	-	-	-
Hip	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ -VE	-	-	✓ ↑ +VE	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ER	-	-	-	-	-	-	-	-	-
Knee	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑FLEX	-	-	-	-	-	-	-	✓ ↑ -VE	✓ ↑ -VE	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	✓ ↑ADD	-	✓ ↑ADD	-	-	-	-	-	✓ ↓ -VE	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ER	-	-	-	-	-	-	-	-	-	-	-	-

✓ Depicts significant difference. IC = initial contact, IR = internal rotation, ER = external rotation, LAT FLEX = Lateral flexion, ABD = Abduction, ADD = Adduction, DF = dorsiflexion, PF = plantarflexion, FLEX = flexion, -VE = negative, +ve = positive

Table 9.3 Unaffected - Matched control comparisons for angular displacement, angular velocity and angular acceleration between movements

		ANGULAR DISPLACEMENT									ANGULAR VELOCITY									ANGULAR ACCELERATION								
		Walk			Land			Cut			Walk			Land			Cut			Walk			Land			Cut		
		Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post
Forefoot-hindfoot	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Y	-	✓ ↓INV	-	-	-	-	-	-	-	-	-	✓ ↓EV	-	-	-	-	✓ ↑INV	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓ABD	✓ ↑ADD	-	-	-	-	-	-	-	-	-
Forefoot-tibia	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	✓ ↓INV	-	-	-	-	-	-	-	-	✓ ↑INV	-	-	-	-	-	-	-	-	-	-
	Z	-	✓ ↓IR	-	-	-	-	-	-	-	✓ ↓ER	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓-VE	✓ ↓-VE
Hindfoot-tibia	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑PF	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓ER	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑+VE	-
Trunk	X	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑EXT	-	✓ ↑FLEX	-	-	-	-	-	-	✓ ↑-VE	✓ ↑-VE	✓ ↑+VE	✓ ↑+VE	-	-
	Y	✓ ↓ADD	-	✓ ↓ADD	-	-	-	-	-	-	-	-	-	-	-	✓ ↓ABD ↑ADD	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ER	✓ ↑ER	-	-	-	-	-	-	-	-	✓ ↑+VE ↑-VE	-	✓ ↑+VE
Hip	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑FLEX	✓ ↑EXT	-	-	-	-	-	-	-	-	-	✓ ↑+VE
	Y	-	-	-	-	-	✓ ↓ADD	-	-	-	-	-	-	-	-	-	✓ ↑ABD	-	-	-	-	-	-	-	-	✓ ↑+VE ↑-VE	-	✓ ↑+VE
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓IR	-	-	-	-	-	-	-	✓ ↑-VE	-	-	-	-	-
Knee	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑FLEX	-	✓ ↑EXT	-	-	-	-	-	-	-	-	-	✓ ↑-VE
	Y	-	-	-	-	-	-	-	-	-	-	✓ ↓ADD	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-

✓ Depicts significant difference. IC = initial contact, IR = internal rotation, ER = external rotation, LAT FLEX = Lateral flexion, ABD = Abduction, ADD = Adduction, DF = dorsiflexion, PF = plantarflexion, FLEX = flexion, -VE = negative, +ve = positive

Table 9.4 CAI group affected - unaffected comparisons for angular displacement, angular velocity and angular acceleration between movements

		ANGULAR DISPLACEMENT									ANGULAR VELOCITY									ANGULAR ACCELERATION								
		Walk			Land			Cut			Walk			Land			Cut			Walk			Land			Cut		
		Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post	Pre	IC	Post
Forefoot-hindfoot	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓PF	-	✓ ↓DF	-	-	-	-	-	-	-	✓ ↓+VE	✓ ↓-VE
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓EV	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
Forefoot-tibia	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓PF	-	-	-	-	-	-	-	-	✓ ↓+VE ↓-VE	-	✓ ↓-VE
	Y	-	✓ ↑INV	✓ ↑INV ↓EV	-	-	-	-	-	-	-	-	✓ ↓INV	-	-	-	✓ ↓EV	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓+VE	-	-	-	-	-	✓ ↑-VE
Hindfoot-tibia	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓ER	-	-	-	-	-	-	-	-	-	-	-
Trunk	X	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑FLEX	-	-	-	-	-	-	✓ ↑-VE	✓ ↓-VE	-	-	-	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ABD ↓ADD	-	-	-	-	-	✓ ↓-VE	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	✓ ↑ER	-	-	✓ ↑IR	✓ ↓ER	-	-	-	-	✓ ↑-VE	-	-	-	✓ ↓+VE	-	-
Hip	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↓FLEX	-	-	-	-	-	-	-	-	✓ ↓-VE
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
Knee	X	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	✓ ↑FLEX	✓ ↓EXT	-	-	-	-	-	-	✓ ↓+VE	-	-
	Y	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-
	Z	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-	-

✓ Depicts significant difference. IC = initial contact, IR = internal rotation, ER = external rotation, LAT FLEX = Lateral flexion, ABD = Abduction, ADD = Adduction, DF = dorsiflexion, PF = plantarflexion, FLEX = flexion, -VE = negative, +ve = positive

9.4.2.1 Clinical Implications

The findings of this research suggest that individual movements should be implemented with careful movement analysis during rehabilitation. It is also important that rehabilitation is performed bilaterally as clear differences have also been observed in the contralateral limb. Lastly, interventions that target the reduction in FFTBA internal rotation may be beneficial.

9.4.3 Regression Analysis

Aim - 'To identify whether it is possible to predict scores on the identification of functional ankle instability questionnaire from the identified variables of significant differences'

The final aim of this thesis was to see whether the variables identified throughout this analysis could be used as predictors for the score on the IdFAI questionnaire, a tool commonly used to determine whether someone is classed as having CAI. This questionnaire uses a cut-off score of 11 or more as instability (Gribble et al., 2013). However, this aimed to determine initial correlations between the kinematic variables and IdFAI score and whether the variables collected may provide a more objective measure of instability. The summary table below provides an overview of each model created from each movement (Table 9.5). Of the three movements, the model with the highest predictive value was the walking model using the FFTBA transverse internal rotation angular displacement 100 ms pre HS, FFTBA transverse positive angular acceleration 100 ms pre HS and trunk lateral flexion displacement from HS to 200 ms post HS which accounts for 49.6% of the variance observed. Further development of this topic may help to provide a valid objective measure for use in a clinical setting to identify those with CAI. Research following an initial ankle sprain may help to identify key differences in full body and multi-segmental foot kinematics providing an invaluable indicator of those susceptible to recurrent sprains. As previously alluded to, lateral ankle sprains account for approximately one-third of the total costs of sports injuries (Kerkhoffs et al., 2012) and long term development of osteoarthritis is probable due to abnormal kinematics (Valderrabano et al., 2006). Early screening methods to identify 'at risk' individuals are critical to improve the quality of life of individuals. The variables suggested within this thesis may provide valuable information for the development of this strategy.

Table 9.5 Regression models from the walking, landing and cutting manoeuvre

	Model	Equation	R	R^2	F	*p* value
Walking	Model 1	IdFAI Score = (-0.925 x FFTBA transverse internal rotation displacement 100 HS) + 16.221 (± 8.479)	0.482	0.232	10.285	0.003
	Model 2	IdFAI Score = (-1.074 x FFTBA transverse internal rotation displacement 100 HS) + (0.542 x FFTBA transverse +ve acceleration 100 HS) + 9.805 (± 7.480)	0.648	0.420	10.683	0.003
	Model 3	IdFAI Score = (-1.015 x FFTBA transverse internal rotation displacement 100 HS) + (0.416 x FFTBA transverse +ve acceleration 100 HS) + (1.457 x Trunk lateral flexion displacement HS 200) + 12.318 (± 7.078)	0.705	0.496	4.858	0.0035
Single Leg Landing	Model 1	IdFAI Score = (-0.181 x Knee sagittal min acceleration IC-200) + 2.415 (± 8.268)	0.520	0.270	12.576	0.001
	Model 2	IdFAI Score = (-0.130 x Knee sagittal min acceleration IC-200) + (0.115 x Knee frontal adduction velocity 200-IC) - 2.568 (± 7.909)	0.593	0.352	4.152	0.050
Cutting	Model 1	IdFAI Score = (0.152 x FFTBA transverse acceleration ICVALUE) + 15.480 (± 8.435)	0.490	0.240	10.742	0.002
	Model 2	IdFAI Score = (0.141 x FFTBA transverse acceleration ICVALUE) + (0.081 x FFTBA frontal Eversion velocity 200-IC) + 19.654 (± 7.857)	0.600	0.360	6.191	0.018
	Model 3	IdFAI Score = (0.115 x FFTBA transverse acceleration ICVALUE) + (0.073 x FFTBA frontal Eversion velocity 200-IC) + (0.145 x FFHFA frontal acceleration ICVALUE) + 19.029 (± 7.445)	0.666	0.443	4.756	0.037

9.4.3.1 Clinical Implications

Further research may be necessary to determine whether the variables outlined within each of the regression models can be used to increase the specificity and sensitivity of tests to confirm ankle instability. Several variables observed in this study were highly correlated to the score on the IdFAI questionnaire suggesting kinematic variables may be a useful predictor of the scores on the IdFAI questionnaire.

9.5 Common Concepts Throughout the Thesis

Throughout Chapter 6, 7 and 8 several significant differences were observed between the matched control and the affected limb of the instability group. Several significant differences were also observed between the unaffected limb and the matched control and the affected limb of the instability group. This suggests that this limb cannot be used within research as the control limb. It remains unclear whether these observed differences were predisposing factors to the initial ankle sprain or whether these adaptations occurred because of the ankle sprain. Clinically these findings may be of importance to therapists working with individuals after an initial sprain and with ankle instability. It highlights the need for exercises to be analysed and closely monitored to ensure they are performed as prescribed. It also appears necessary to ensure that exercises not only address the affected limb but also the unaffected limb.

Future research should continue to investigate not just angular displacement but also angular velocities and accelerations as this research shows that they may be pertinent in those with ankle instability.

A key finding across this thesis is the decreased internal rotation angular displacement of the forefoot in relation to the tibia prior to IC. Further research should investigate this finding within different movements. Cadaveric analysis may be particularly beneficial when investigating how this may impact the dynamic restraint mechanisms for lateral ankle sprains.

9.6 Thesis Development

The initial proposal for this research study was to further investigate simulated ankle sprains using a tilt platform device. On further research of this area, a number of key limitations were identified including the lack of applicability of a bilateral stance (Knight & Weimar, 2011a) and the potential pre-activation (Hopkins et al., 2007) thought to occur in preparation for the tilt. Added to this, there is a paucity of research available that fully understands the mechanism of a lateral ankle sprain (Fong et al., 2012; Kristianslund et al., 2011). Pilot testing found large deviations in tilt velocity between participants of different masses. As such it was thought more beneficial to better understand the biomechanics adopted by individuals with ankle instability and healthy control participant. This being carried out to identify potential movement strategies that could be targeted for clinical interventions.

Research has reported a high incidence of lateral ankle sprains sustained in a general population (Fong et al., 2007; Gribble et al., 2016). Furthermore, regular complaints of giving way sensations on uneven and level surfaces during walking have also been identified (Wright et al., 2013a). Therefore, the first study of this thesis (Chapter 6) aimed to identify potential differences in kinematics and muscle activation in individuals with CAI during walking. Due to the high incidence of non-contact lateral ankle sprains previously reported during landings (Woods et al., 2003), Chapter 7 aimed to investigate kinematics and muscle activation in individuals with CAI during single-leg landing manoeuvres. Lastly, due to the high incidence of ankle sprains sustained during twisting and turning motions (Woods et al., 2003), Chapter 8 investigated kinematics and muscle activation during cutting manoeuvres.

Following statistical parametric mapping and discrete variable analysis, regression analysis was performed to determine whether the score on the IdFAI questionnaire could be predicted from the analysed variables to provide a more objective measure of ankle instability.

9.7 Limitations

Chronic ankle instability was determined using a questionnaire which may evoke some criticism, however, this is recommended practice in line with the guidelines set out by the IAC (Gribble et al., 2013). This also provided the rationale for the regression analysis adopted for each movement to see if the variables observed in the present research may be a valid predictor of ankle instability rather than the use of a cut-off scale.

Another possible limitation is the absence of force plates used within this thesis. This was outside of the scope of this research, however, does leave some speculation regarding the possible reasons for the observed results. In a clinical setting 2D kinematics are often observed however the cost and time constraints of integrating force plates is significant. This research highlights key areas of differences between groups. Future research could look to see whether 2D analysis may produce similar results to the present study for use in a clinical setting.

One possible limitation of this research is that each manoeuvre was conducted barefoot. Although this lacks replicability in most sporting contexts, it was felt most appropriate to avoid modification of kinematics (Morio et al., 2009; Scott et al., 2012) with the use of a standardised shod condition or by adding additional variability to the data with the use of participants' habitual shoes. Similarly, no measures of foot structure were obtained which may account for some of the variations observed.

Although a multi-segmental foot model has been used in this research it is important to note that a single segment trunk model was used. To the author's knowledge this is the first study to investigate trunk kinematics during dynamic activities with individuals with ankle instability as such a simplified model was used for these preliminary investigations. Future studies may wish to further the results of this thesis by investigating with a multi-segmental trunk model.

Within this research, the initial mechanism of injury was not reported by participants. This is a potential limitation of the IdFAI questionnaire as this may have provided vital information firstly on whether their initial sprain was a contact or non-contact mechanism, but this may also lead us to better understand which structures may have been injured.

9.8 Future Research

Future research should aim to address the following areas:

- Incorporating kinetic analysis to compare between those with CAI and healthy matched controls during the previously analysed movements. This may provide further understanding with ground reaction force data as well as joint moments and analysis of the centre of gravity which may provide greater insight into the observed findings of this thesis
- Adopting a prospective research method to identify whether the findings of this study may be early predictors of lateral ankle sprains or whether these are preventative measures adopted following an initial sprain

- Implementation of injury preventative strategies based on some of the findings outlined in the current thesis
- Identification of a tool to identify 'at risk' individuals that can be targeted following an initial ankle sprain. The results presented in this study may provide a basis for some of this further work.

Chapter 10.0 Conclusion

10.1 General Conclusion

Decreased FFTBA internal rotation angular displacement has been found to occur prior to IC in the affected limb of the CAI group when compared to a matched control in all three of the observed movements. As such this variable should be of interest for future research and formulation of preventative strategies. Key differences have also been observed between the affected limb of the CAI group and a healthy matched control but also when compared to the unaffected limb and when comparing the unaffected limb to a healthy matched control. This suggests that bilateral movement analysis and education is critical in the rehabilitation of athletes and the general population from lateral ankle sprains. Lastly, excluding FFTBA internal rotation, differences observed between groups seem to be individual to each movement and as such intervention strategies adopting a one size fits all approach may be ineffective for targeting each of the movements that are common in the mechanisms of lateral ankle sprain.

In summation, this exploratory research has identified some key differences both in individual movements but some also across movements. These differences are identified in not just foot and ankle kinematics but also higher up the kinetic chain in the trunk, hip and knee. Differences are also highlighted in the contralateral limb. These findings may therefore be used in the development of intervention and rehabilitation strategies and in the development of screening strategies. This could help to aid in the prevention of CAI and, improve the quality of life for those struggling with the condition.

Chapter 11.0 References

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Chapter 12.0 Appendices

Appendix A Consent Form

UNIVERSITY OF HERTFORDSHIRE
ETHICS COMMITTEE FOR STUDIES INVOLVING THE USE OF HUMAN PARTICIPANTS
('ETHICS COMMITTEE')

FORM EC3
CONSENT FORM FOR STUDIES INVOLVING HUMAN PARTICIPANTS

I, the undersigned [please give your name here, in BLOCK CAPITALS]

……………………………………………………………………………………………………….…

of [please give contact details here, sufficient to enable the investigator to get in touch with you, such as a postal or email address]

….……………………………………………………………………………………………………

hereby freely agree to take part in the study entitled [insert name of study here]

……………………………………………………………………………………………………..

1 I confirm that I have been given a Participant Information Sheet (a copy of which is attached to this form) giving particulars of the study, including its aim(s), methods and design, the names and contact details of key people and, as appropriate, the risks and potential benefits, and any plans for follow-up studies that might involve further approaches to participants. I have been given details of my involvement in the study. I have been told that in the event of any significant change to the aim(s) or design of the study I will be informed and asked to renew my consent to participate in it.

2 I have been assured that I may withdraw from the study at any time without disadvantage or having to give a reason.

3 In giving my consent to participate in this study, I understand that voice, video or photo-recording will take place.

4 I have been given information about the risks of my suffering harm or adverse effects. I have been told about the aftercare and support that will be offered to me in the event of this happening, and I have been assured that all such aftercare or support would be provided at no cost to myself.

5 I have been told how information relating to me (data obtained in the course of the study, and data provided by me about myself) will be handled: how it will be kept secure, who will have access to it, and how it will or may be used.

6 I understand that my participation in this study may reveal findings that could indicate that I might require medical advice. In that event, I will be informed and advised to consult my GP. If, during the study, evidence comes to light that I may have a pre-existing medical condition that may put others at risk, I understand that the University will refer me to the appropriate authorities and that I will not be allowed to take any further part in the study.

7 I understand that if there is any revelation of unlawful activity or any indication of non-medical circumstances that would or has put others at risk, the University may refer the matter to the appropriate authorities.

8 I have been told that I may at some time in the future be contacted again in connection with this or another study.

Signature of participant……………………………………..….Date………………………..

Signature of (principal) investigator……………………………………………………Date………………………

Name of (principal) investigator [in BLOCK CAPITALS please]……………………………………………………………

Appendix B Health Screen

UNIVERSITY OF HERTFORDSHIRE **Researcher:**
SCHOOL OF LIFE SCIENCE
HEALTH SCREEN 1
Title of Study:
Subject Name:

It is important when having volunteered as subject for this study, and having read the briefing sheet for subjects that you answer the following questions. Please do not answer any questions if you consider them intrusive.

Do you suffer from high blood pressure, or any heart problems?

Yes	No

Do you often get dizzy, or do you know that you have low blood pressure?

Yes	No

When and what did you last eat?

Are you under the influence of alcohol or any other psycho-active substance?

Yes	No

Have you had a cold or flu in the last two weeks?

Yes	No

Are you suffering from any musculoskeletal injury?

Yes	No

Are you currently taking any medication (over the counter, or prescription)?

Yes	No

(you do not need to answer "Yes" if you are only taking oral contraceptives, or if you are an asthmatic with an inhaler available)

Have you ever been told that you should not exercise?

Yes	No

Do you feel fully fit, and eager to act as subject?

Yes	No

Is there any reason, not stated above, why you cannot take part as a subject in this practical?

Yes	No

Signature………………………………………………….. Date:

Checked by (Name): Date:

Appendix C Participant information sheet

UNIVERSITY OF HERTFORDSHIRE

ETHICS COMMITTEE FOR STUDIES INVOLVING THE USE OF HUMAN PARTICIPANTS

('ETHICS COMMITTEE')

FORM EC6: PARTICIPANT INFORMATION SHEET

Title of study

Influence of Chronic Ankle Instability on Human Movement: A Three Dimensional Kinematic and Electromyographic Analysis

Introduction

You are being invited to take part in a study. Before you decide whether to do so, it is important that you understand the research that is being done and what your involvement will include. Please take the time to read the following information carefully and discuss it with others if you wish. Do not hesitate to ask us anything that is not clear or for any further information you would like to help you make your decision. Please do take your time to decide whether or not you wish to take part. Thank you for reading this.

What is the purpose of this study?

Ankle sprains are one of the most common sporting injuries. It has been previously stated that in order to improve preventative measures for ankle sprains a better understanding of the mechanism is needed with biomechanical quantities. Few studies have used three-dimensional motion analysis for the study of ankle sprains and those that have, have not commented upon movement above the tibia. This study measure kinematics and muscle activation during dynamic activity.

Do I have to take part?

It is completely up to you whether or not you decide to take part in this study. If you do decide to take part you will be given this information sheet to keep and be asked to sign a consent form. Agreeing to join the study does not mean that you have to complete it. You are free to withdraw at any stage without giving a reason. A decision to withdraw at any time, or a decision not to take part at all, will not affect any treatment/care that you may receive (should this be relevant).

How long will my part in the study take?

If you decide to take part in this study, you will be required to spend an hour in the on-site biomechanics laboratory (G105a) on College Lane.

Am I eligible to take part in the study?

Recreationally active individuals aged 18-35 and participating in sport a minimum of 2 times a week will be recruited to participate in this study.

When should I refuse to take part?

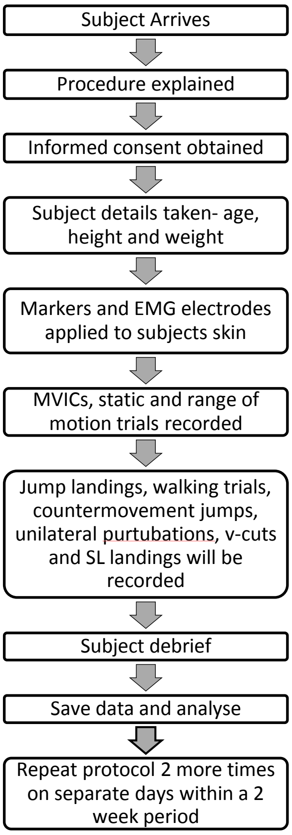
You must not have:

- Existing lower limb injuries
- Prescribed orthotics
- Lower-extremity biomechanical abnormality
- Balance or motion disorder
- History of lower extremity surgery
- Suffered an ankle sprain within 6 months

What will happen to me if I take part?

Subjects will be required to wear cycling shorts and females will be asked to wear a sports bra. On arrival the test procedure will be explained and the subject will be given the opportunity to ask any questions they may have. Permission to continue will then be requested though subjects will be reminded of their right to withdraw from testing without reason at any point. Characteristic measures will be taken consisting of age, gender, height and mass.

Motion analysis data will be recorded using the Owl Digital Real Time 10 Camera System (Motion Analysis, Santa Rosa, California) to track movement trajectories of the reflective markers attached to the subject. A modified Helen Hayes marker set will be used with the oxford foot model to digitise points across the whole body. This will enable analysis of whole body movement as well as detailed analysis of the lower limb. Markers will be attached using hypoallergenic double-sided marker stickers directly to the skin. A static trial and a range of motion trial will then be conducted in order to obtain marker reference points.



Electromyographic data will be recorded using the Biometrics datalog W4X8 Bluetooth unit (Gwent, Wales) to measure muscle activation and muscle onset times. SX230 surface electrodes will be applied to the muscles in accordance with SENIAM guidelines with an interelectrode distance of 20mm. Electrodes will be aligned parallel to the muscle belly for each muscle. Three maximal contractions will be recorded for each muscle.

Subjects will be required to complete jump landings, walking trials, countermovement jumps, unilateral perturbations, cuts and single leg landings for each foot.

Following testing the subject will be debriefed to clarify test data use and asked to report any side effects they are feeling. Data will then be saved and analysed.

What are the possible disadvantages, risks or side effects of taking part?

There is a small possibility that falls or trips may be encountered off the tilt platform or during jump landing but the subject will be shown how the equipment works prior to its use and will experience familiarization to the equipment.

What are the possible benefits of taking part?

Subjects will have their biomechanics analysed and therefore advice can be given on any biomechanical abnormalities that are found during the testing therefore assisting with performance and decreasing the likelihood of injury. Students will also experience the use of the new camera system that has recently been installed.

How will my taking part in this study be kept confidential?

All data will be stored on password protected computers and paper files in locked filing cabinets with your name and personal details removed.

What will happen to the data collected within this study?

Results of the research may be published in a scientific journal and be presented at conferences but it will not be possible to identify individual participants.

Who has reviewed this study?

The research project has been reviewed by the universities ethics committee. The design and suitability has also been discussed with Gerwyn Hughes the principal supervisor for the study. It has also been approved by laboratory manager.

Who can I contact if I have any questions?

If you would like further information or would like to discuss any details personally, please get in touch with me, in writing, or by email:

Lynsey Northeast- G111, CP Snow Building

Email: l.northeast@herts.ac.uk

Although we hope it is not the case, if you have any complaints or concerns about any aspect of the way you have been approached or treated during the course of this study, please write to the University's Secretary and Registrar.

Thank you very much for reading this information and giving consideration to taking part in this study.

Appendix D Identification of Functional Ankle Instability Questionnaire

IDENTIFICATION OF FUNCTIONAL ANKLE INSTABILITY (IdFAI)

Instructions: This form will be used to categorize your ankle stability status. A separate form should be used for the right and left ankles. Please fill out the form completely and if you have any questions, please ask the administrator. Thank you for your participation.

Please carefully read the following statement:
"Giving way" is described as a temporary uncontrollable sensation of instability or rolling over of one's ankle.

I am completing this form for my **RIGHT/LEFT** ankle (circle one).

1.) Approximately how many times have you sprained your ankle? ________

2.) When was the last time you sprained your ankle?

❑Never	❑ > 2 years	❑ 1-2 years	❑ 6-12 months	❑ 1-6 months	❑< 1 month
0	1	2	3	4	5

3.) If you have seen an athletic trainer, physician, or healthcare provider how did he/she categorize your most serious ankle sprain?

❑Have **not** seen someone	❑Mild (Grade I)	❑Moderate (Grade II)	❑Severe (Grade III)
0	1	2	3

4.) If you have ever used crutches, or other device, due to an ankle sprain how long did you use it?

❑Never used a device	❑1-3 days	❑4-7 days	❑1-2 weeks	❑2-3 weeks	❑>3 weeks
0	1	2	3	4	5

5.) When was the last time you had ***"giving way"*** in your ankle?

❑Never	❑> 2 years	❑1-2 years	❑6-12 months	❑ 1-6 months	❑< 1 month
0	1	2	3	4	5

6.) How often does the ***"giving way"*** sensation occur in your ankle?

❑Never	❑Once a year	❑Once a month	❑Once a week	❑Once a day
0	1	2	3	4

7.) Typically when you start to roll over (or 'twist') on your ankle can you stop it?

❑Never rolled over	❑Immediately	❑Sometimes	❑Unable to stop it
0	1	2	3

8.) Following a typical incident of your ankle rolling over, how soon does it return to 'normal'?

❑Never rolled over	❑Immediately	❑ < 1 day	❑1-2 days	❑> 2 days
0	1	2	3	4

9.) During "Activities of daily life" how often does your ankle feel ***UNSTABLE***?

❑Never	❑Once a year	❑Once a month	❑Once a week	❑Once a day
0	1	2	3	4

10.) During "Sport/or recreational activities" how often does your ankle feel ***UNSTABLE***?

❑Never	❑Once a year	❑Once a month	❑Once a week	❑Once a day
0	1	2	3	4

Figure D.1 Identification of functional ankle instability questionnaire (Simon et al., 2012)

Appendix E Study 1 part 1 – Walk SPM

Matched Control vs Affected limb of CAI group

- **Heel strike to Toe off**

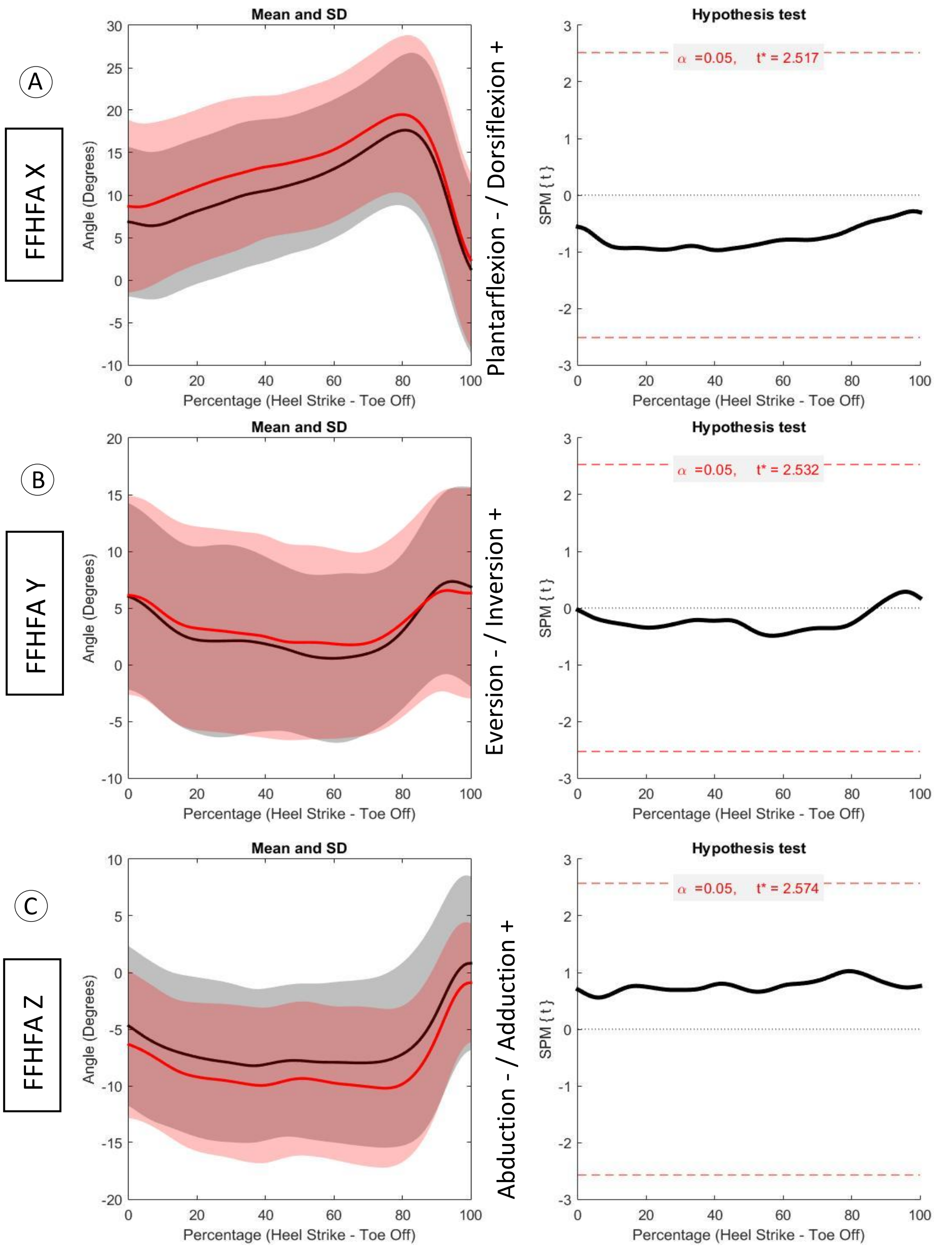


Figure E.1 Walking - Forefoot – Hindfoot angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

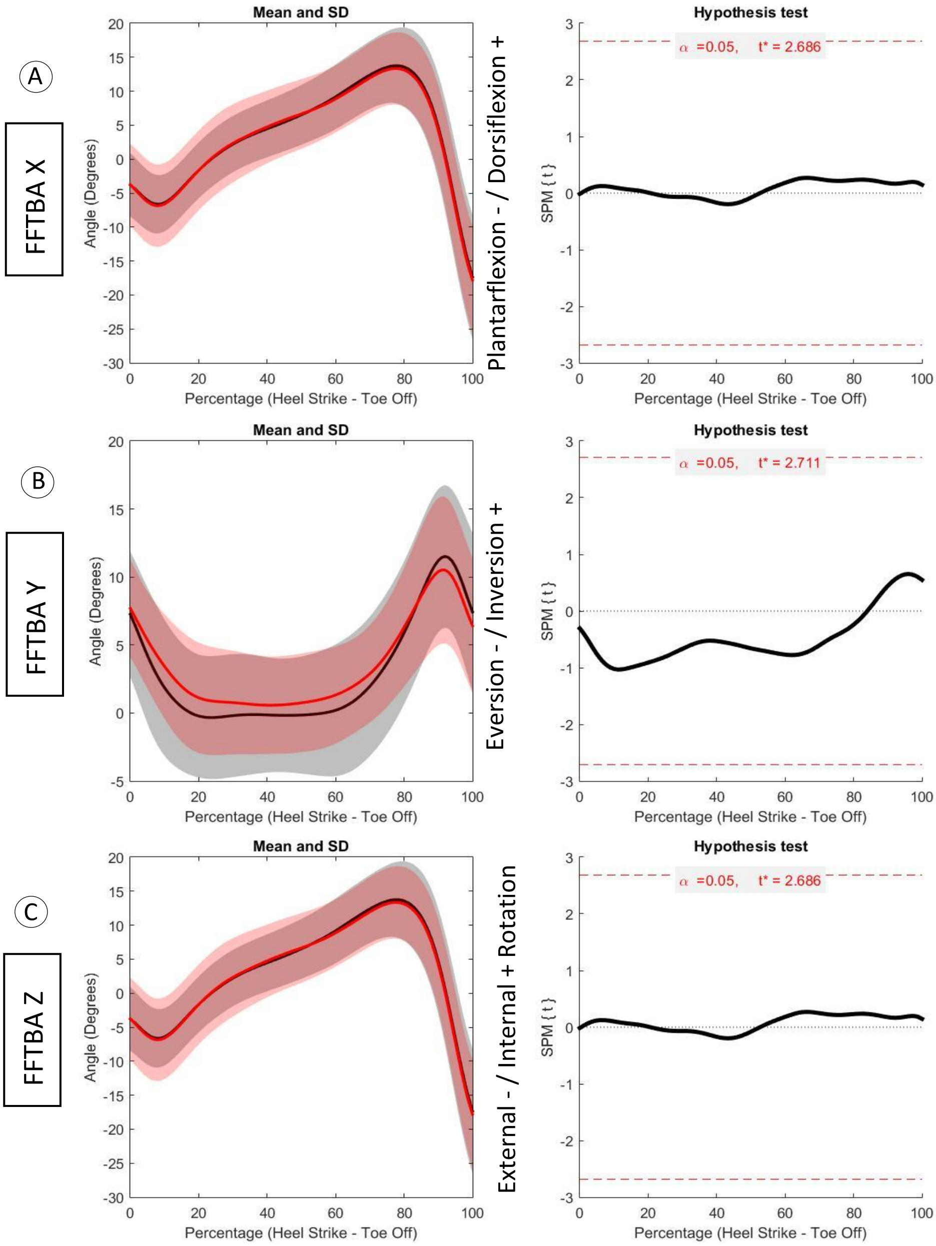


Figure E.2 Walking - Forefoot – tibia angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

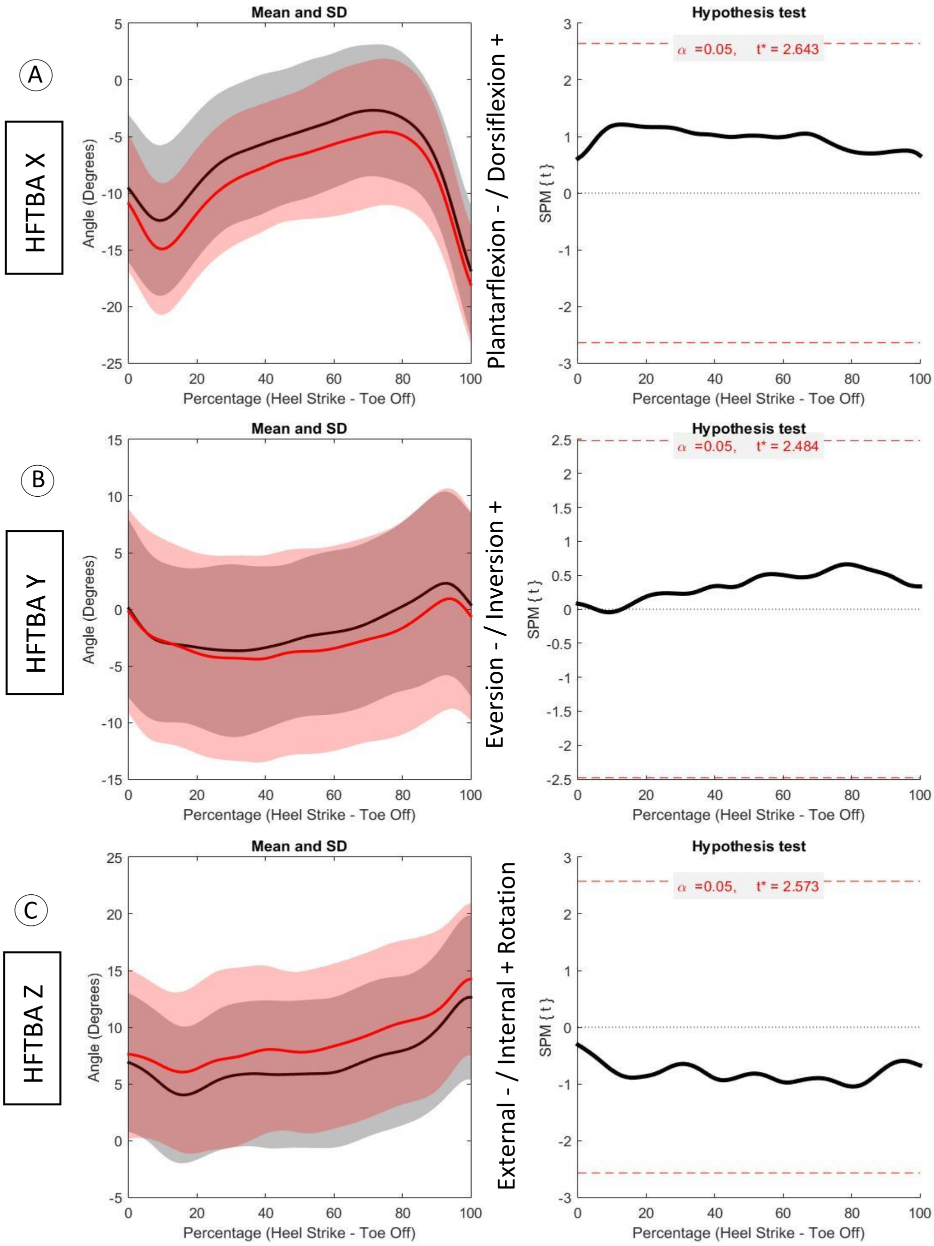


Figure E.3 Walking - Hindfoot – tibia angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

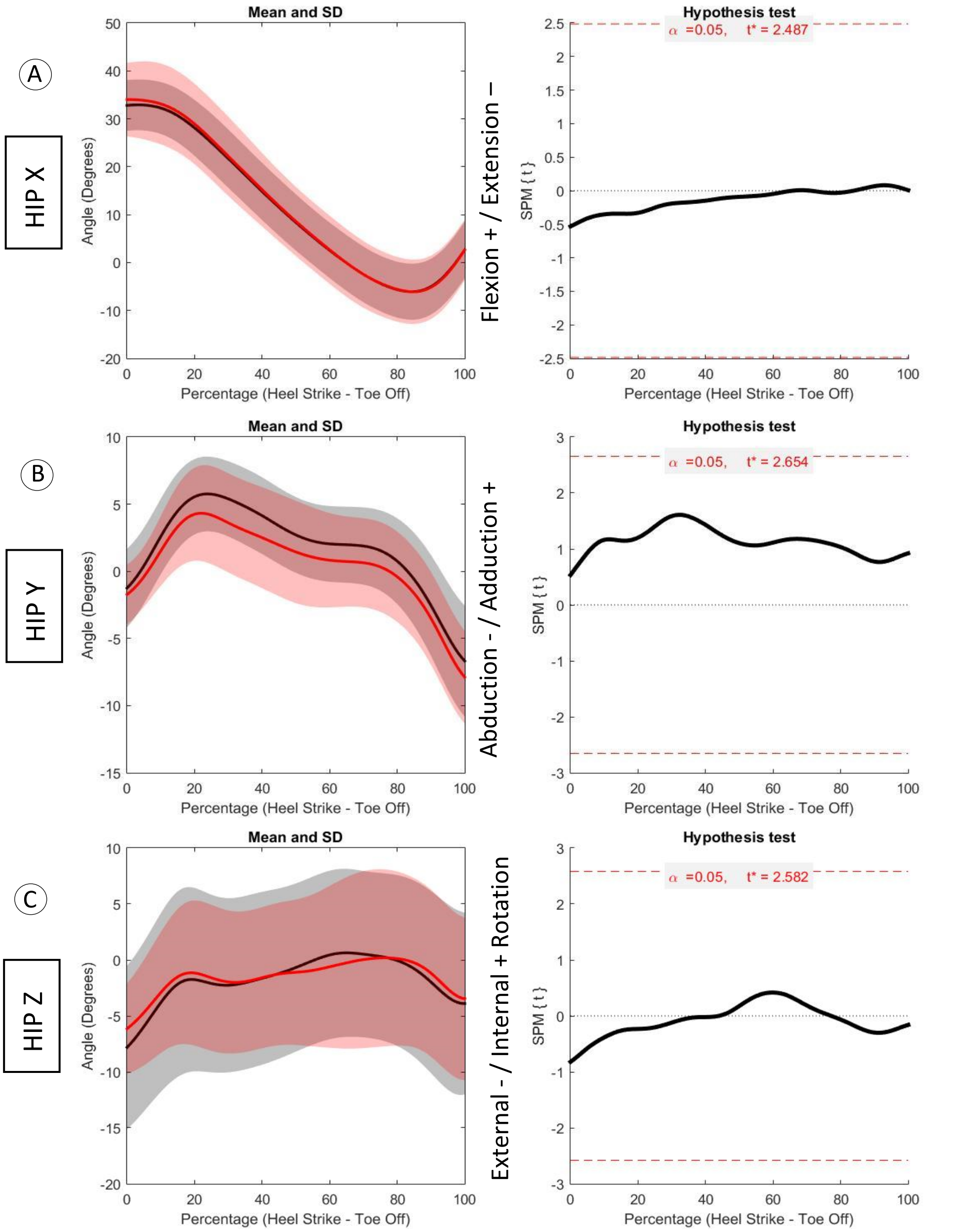


Figure E.4 Walking - Hip angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

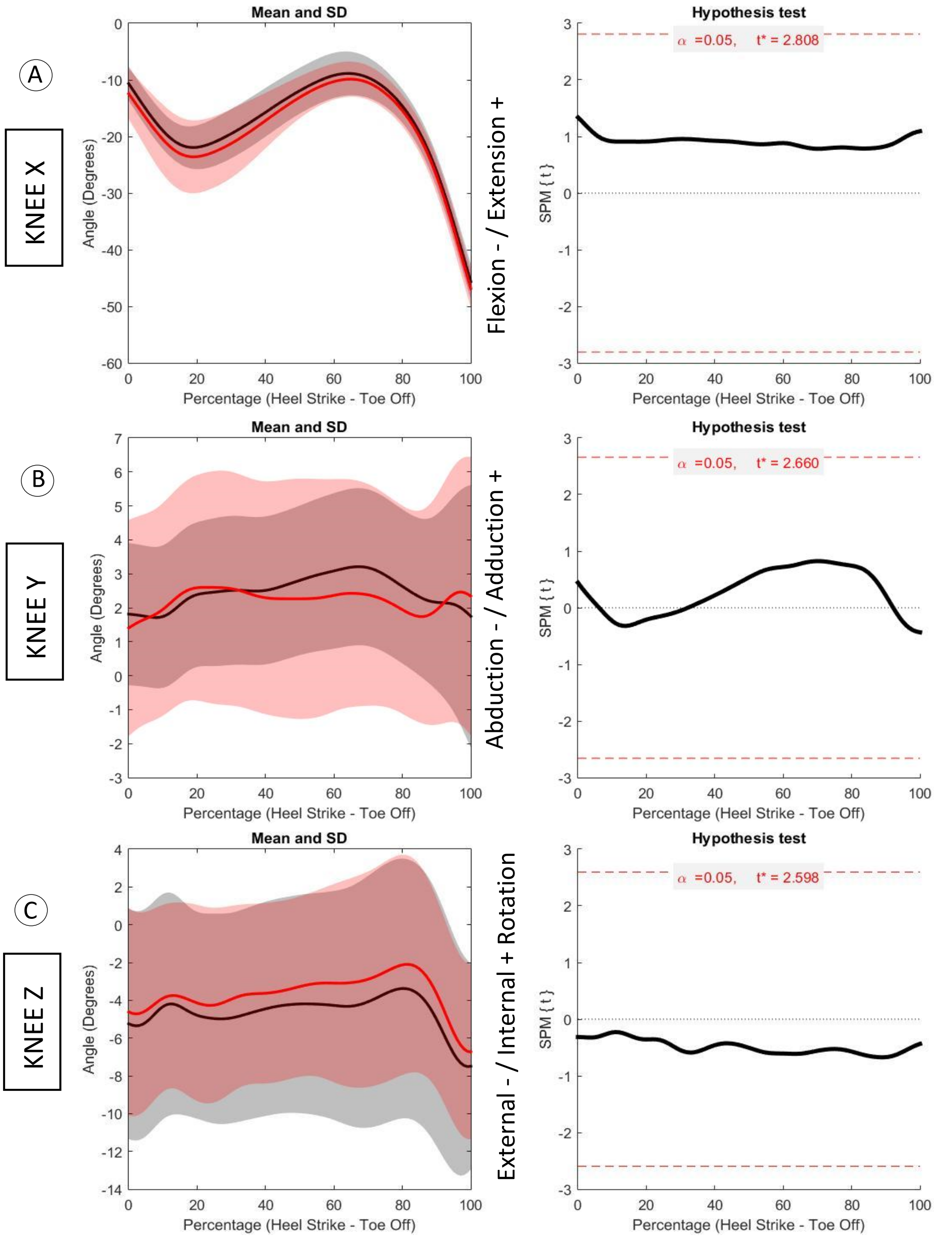


Figure E.5 Walking - Knee angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

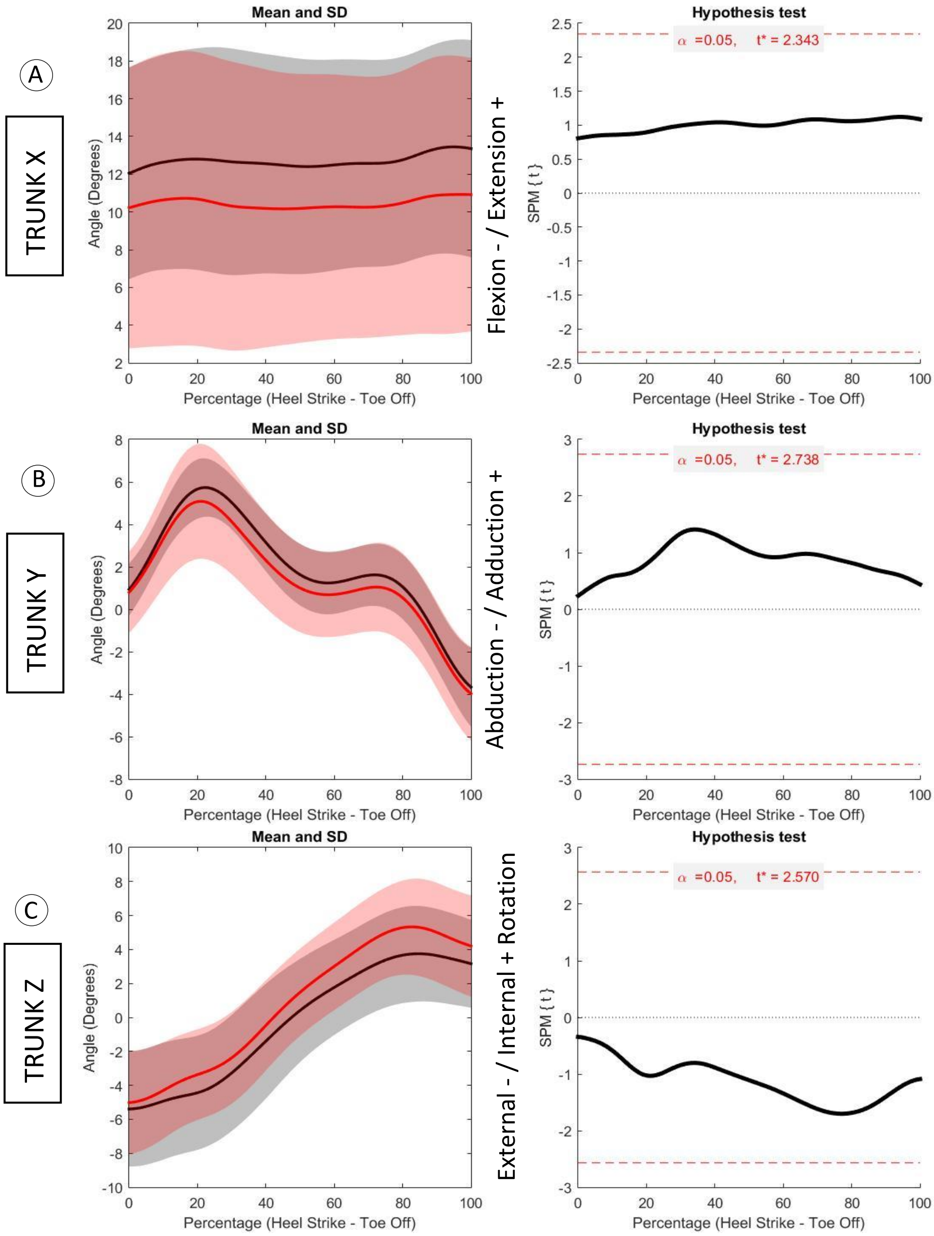


Figure E.6 Walking - Trunk angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

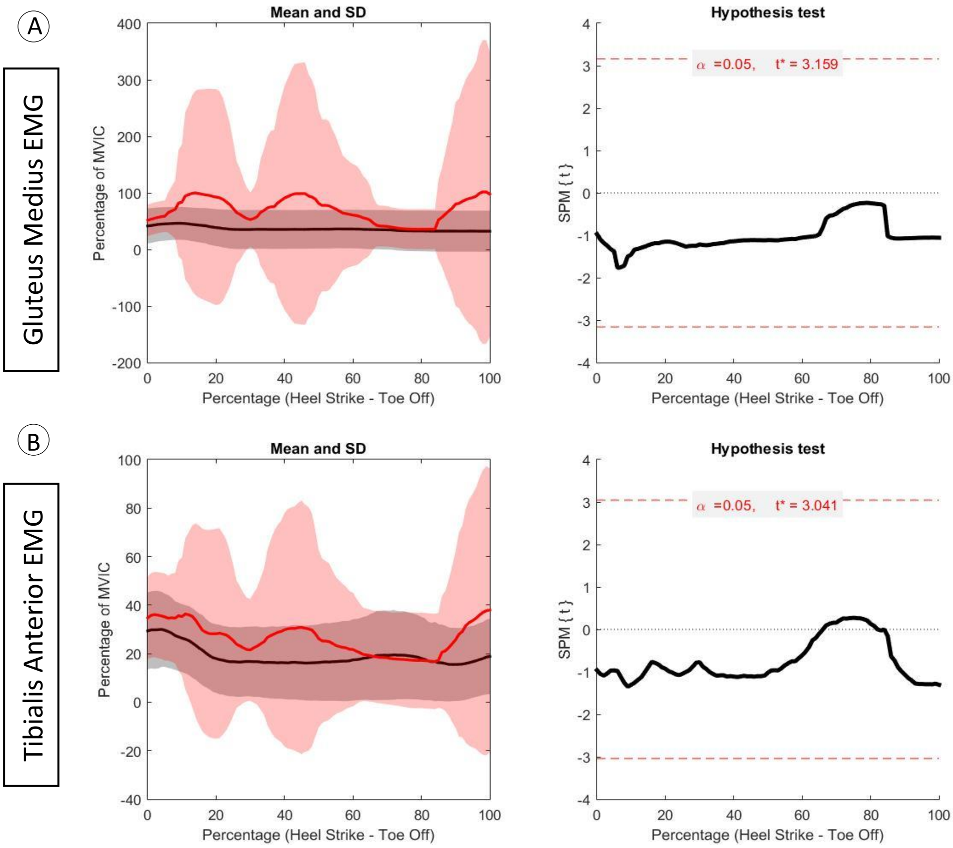


Figure E.7 Walking - Gluteus Medius and Tibialis Anterior muscle activation – heel strike - toe off means and standard deviations (Matched Control -— Affected - —) and *t*-test output.

- **Toe off to Heel strike**

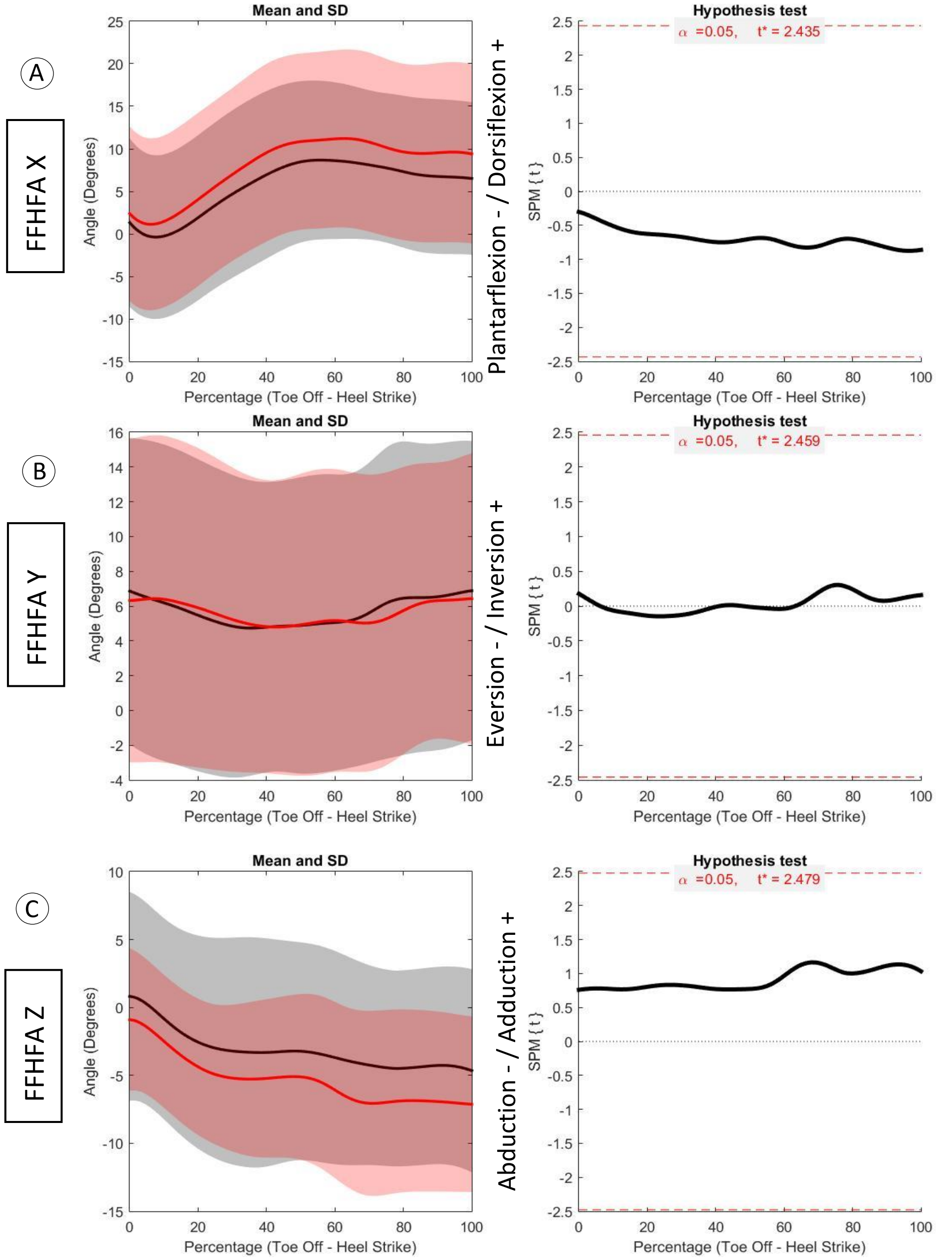


Figure E.8 Walking - Forefoot – Hindfoot angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

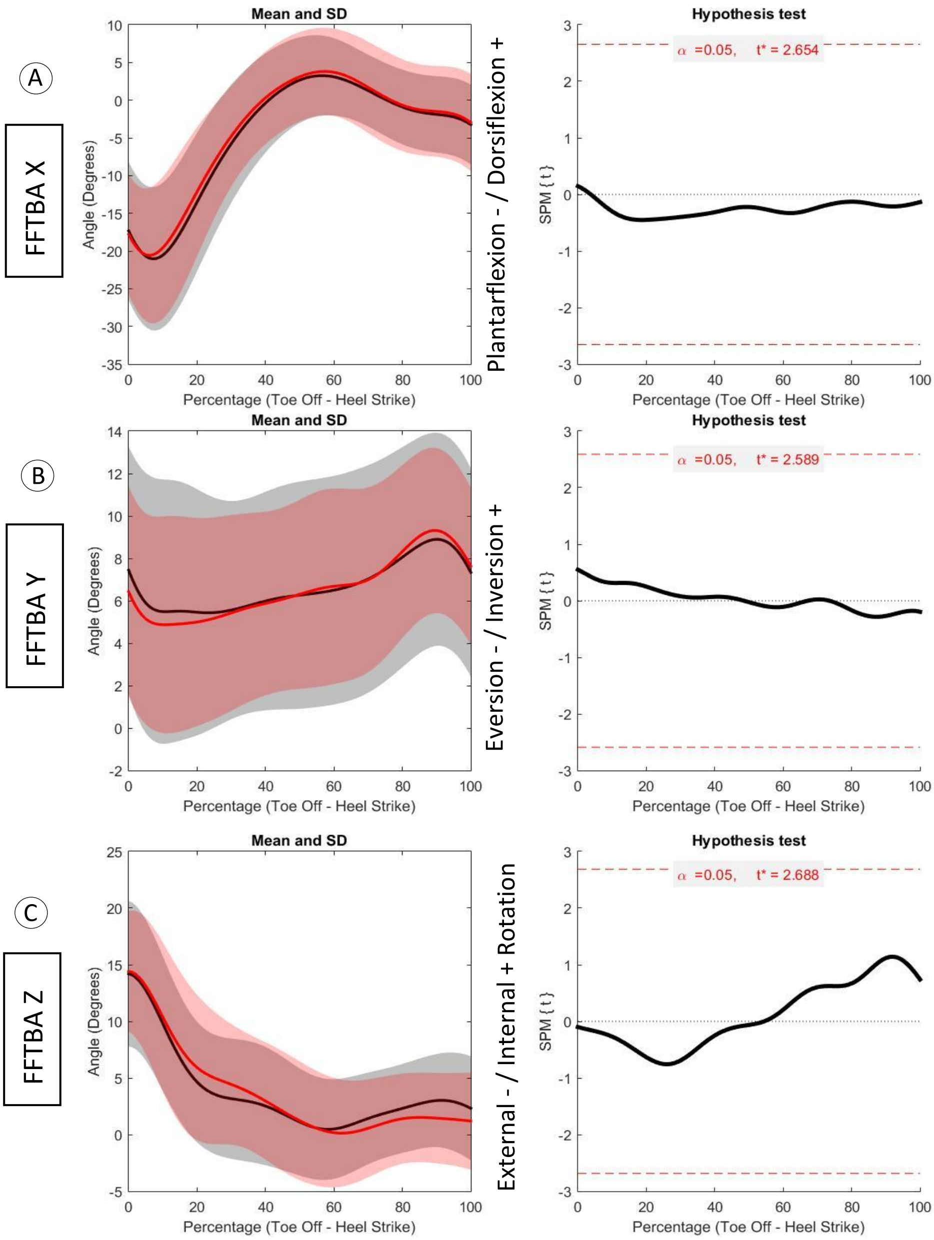


Figure E.9 Walking - Forefoot –Tibia angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

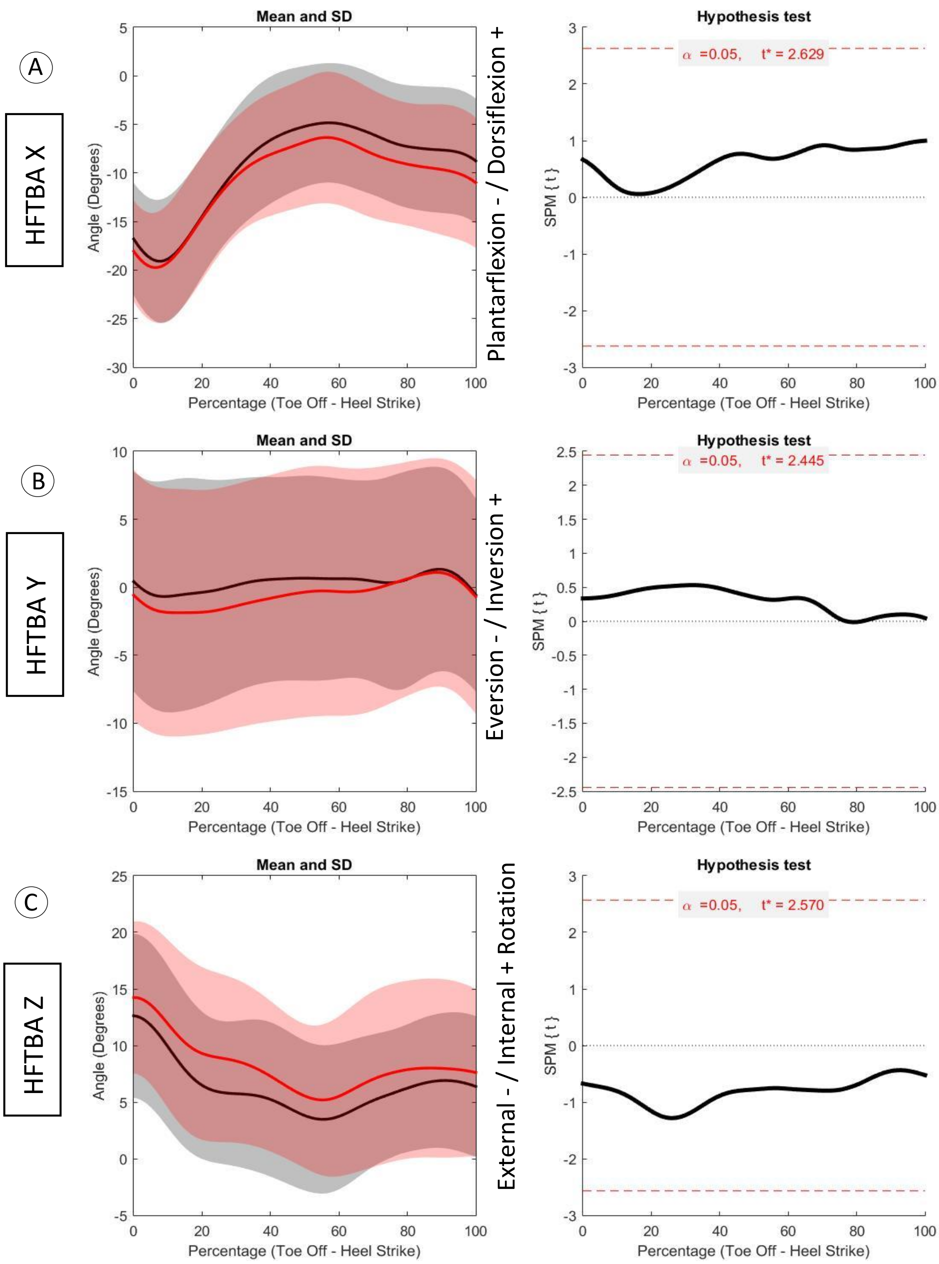


Figure E.10 Walking - Hindfoot-Tibia angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

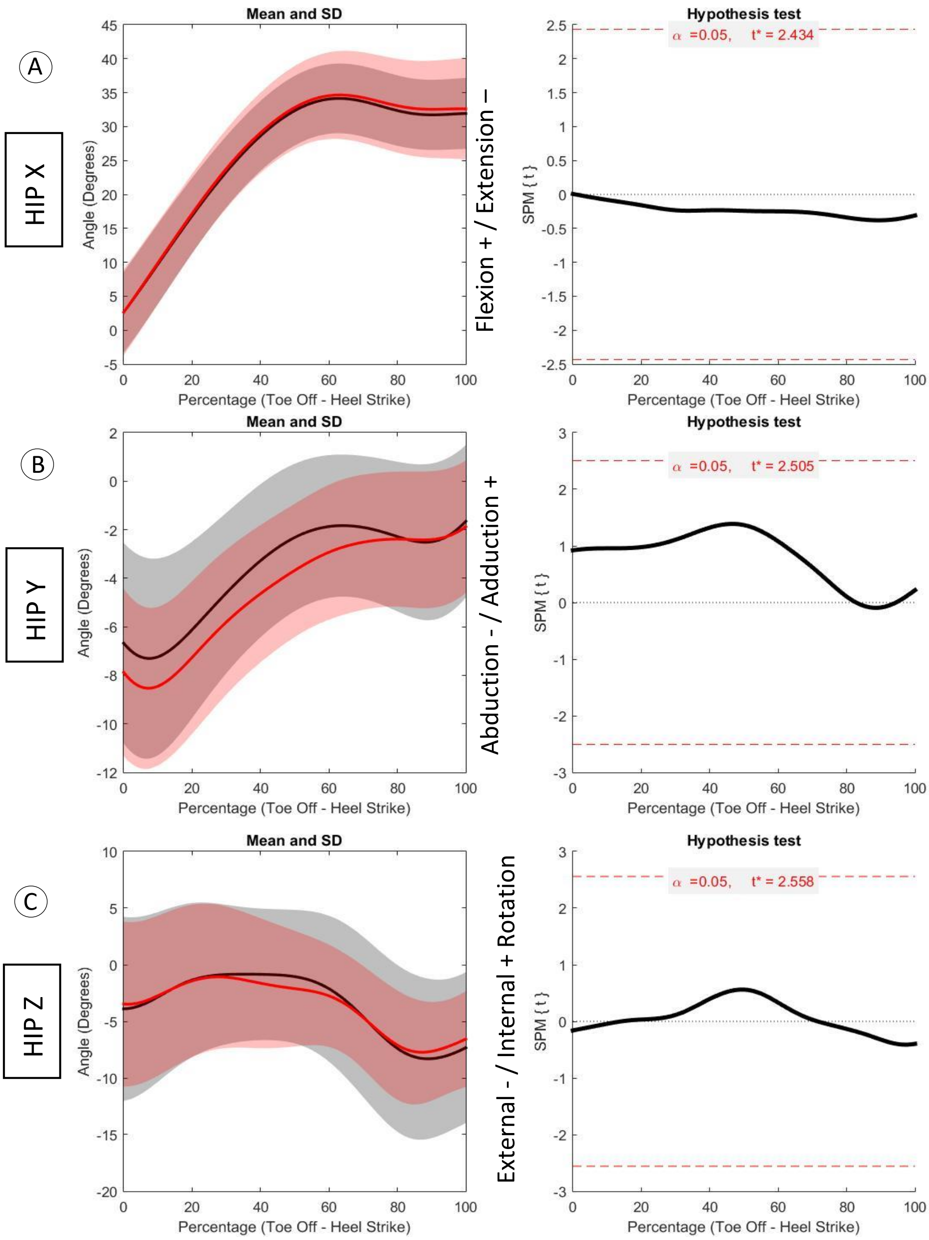


Figure E.11 Walking – Hip angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

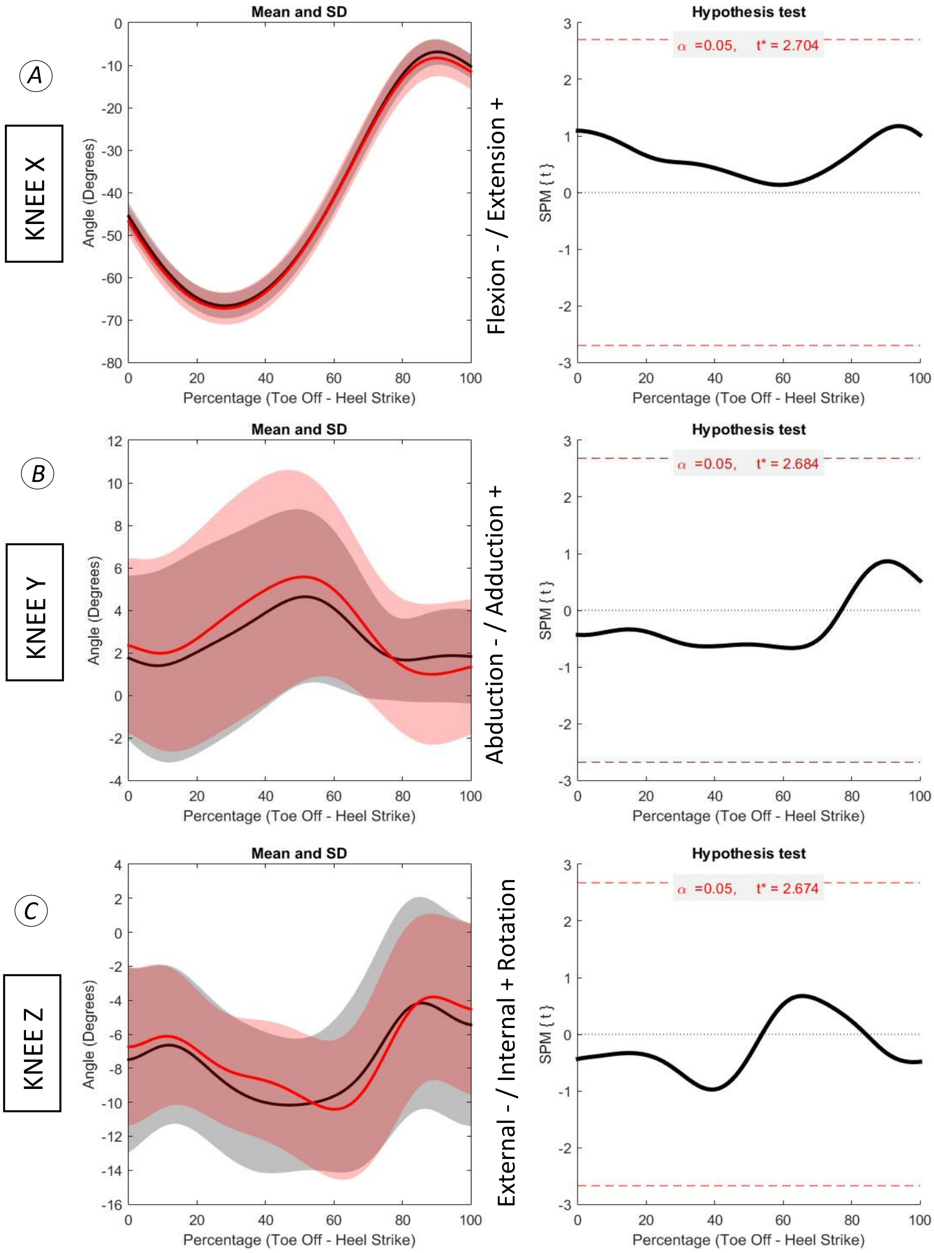


Figure E.12 Walking - Knee angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

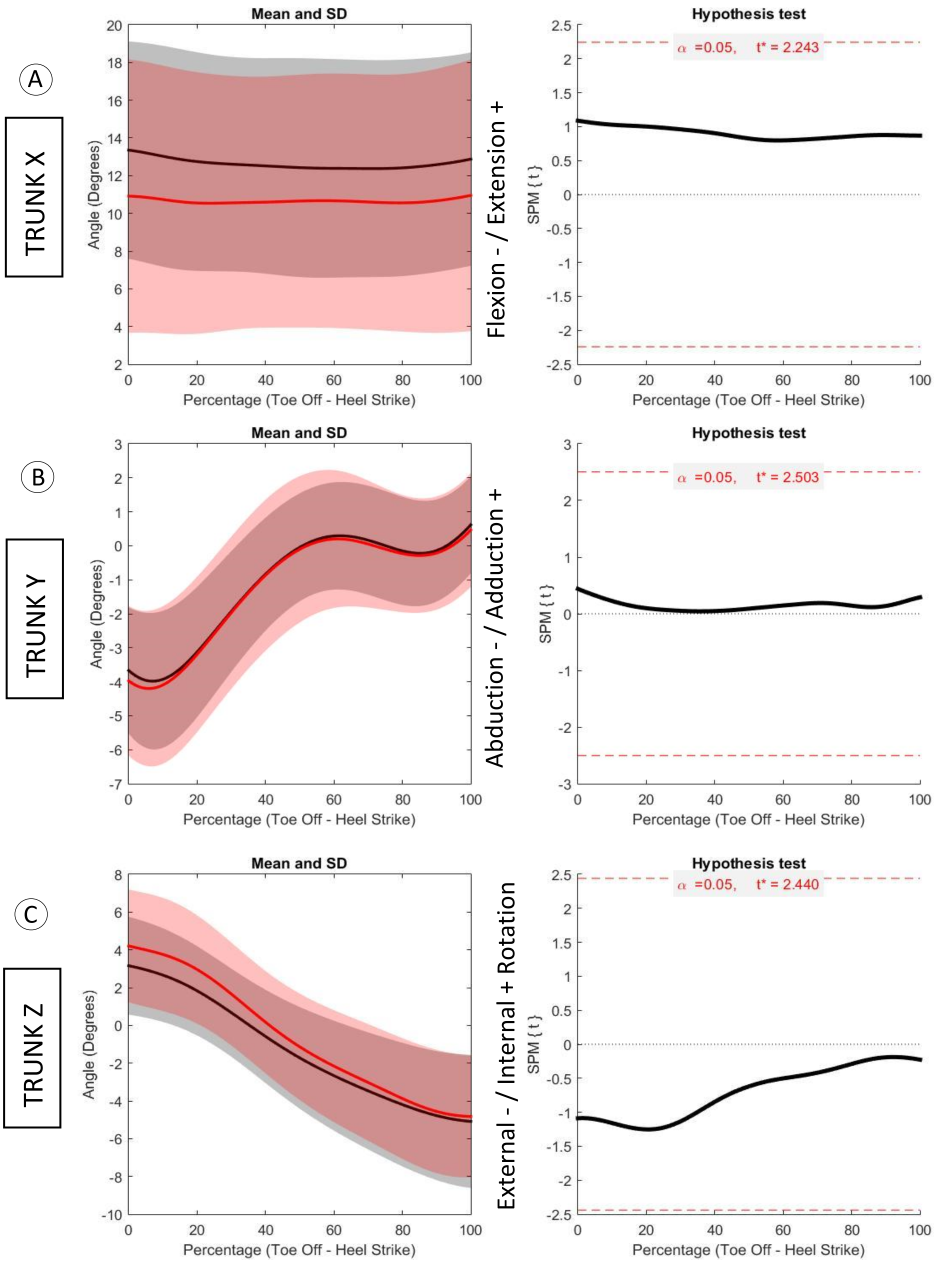


Figure E.13 Walking - Trunk angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

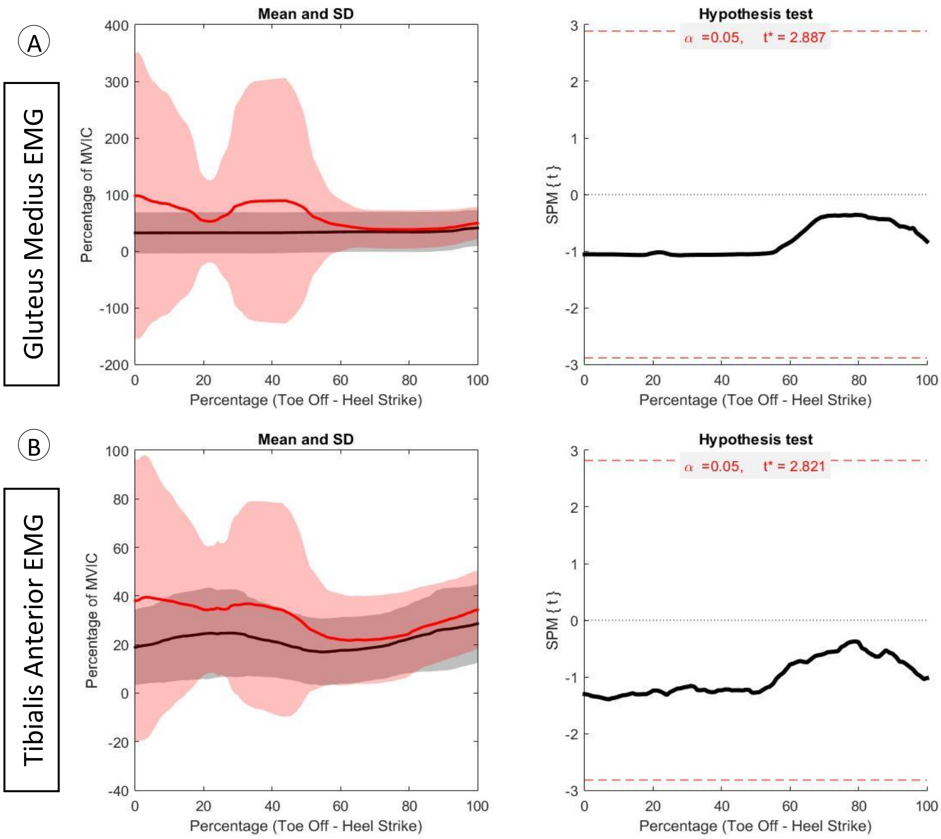


Figure E.14 Walking - Gluteus Medius and Tibialis Anterior muscle activation – toe off - heel strike - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

Unaffected limb vs Affected limb of CAI group

- **Heel strike to Toe off**

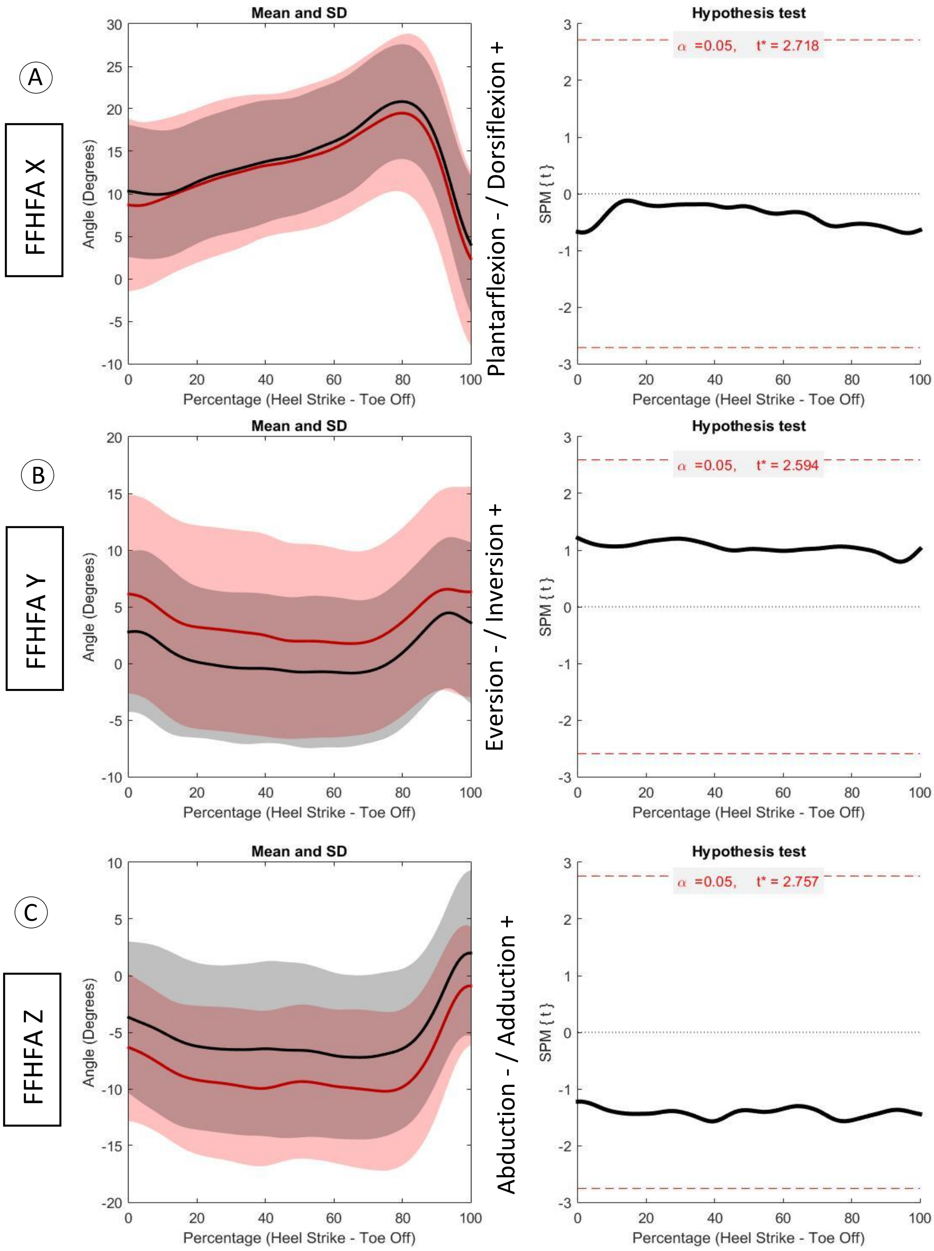


Figure E.15 Walking - Forefoot – Hindfoot angles (x, y, z) Heel strike - Toe off – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

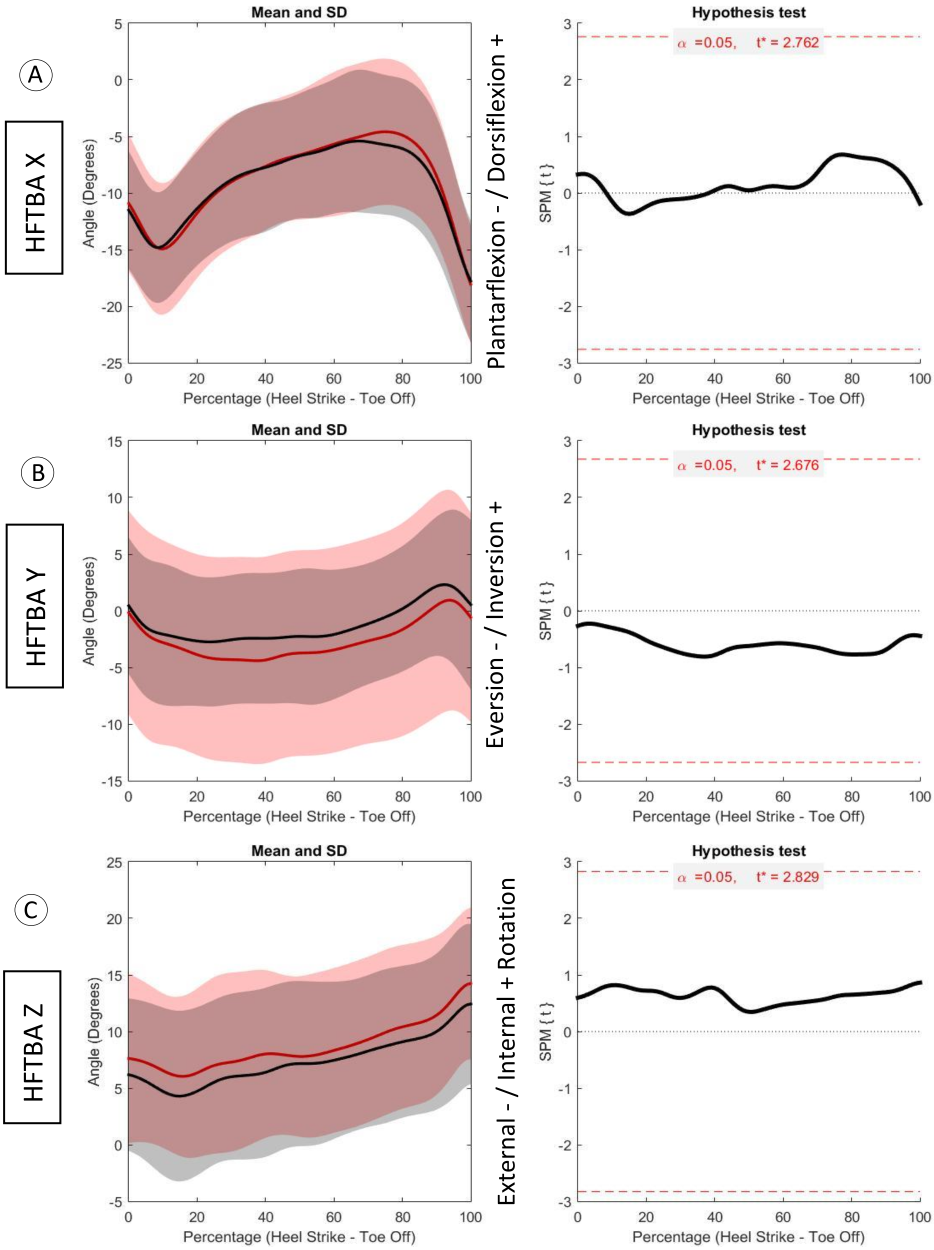


Figure E.16 Walking - Hindfoot – Tibia angles (x, y, z) Heel strike - Toe off – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

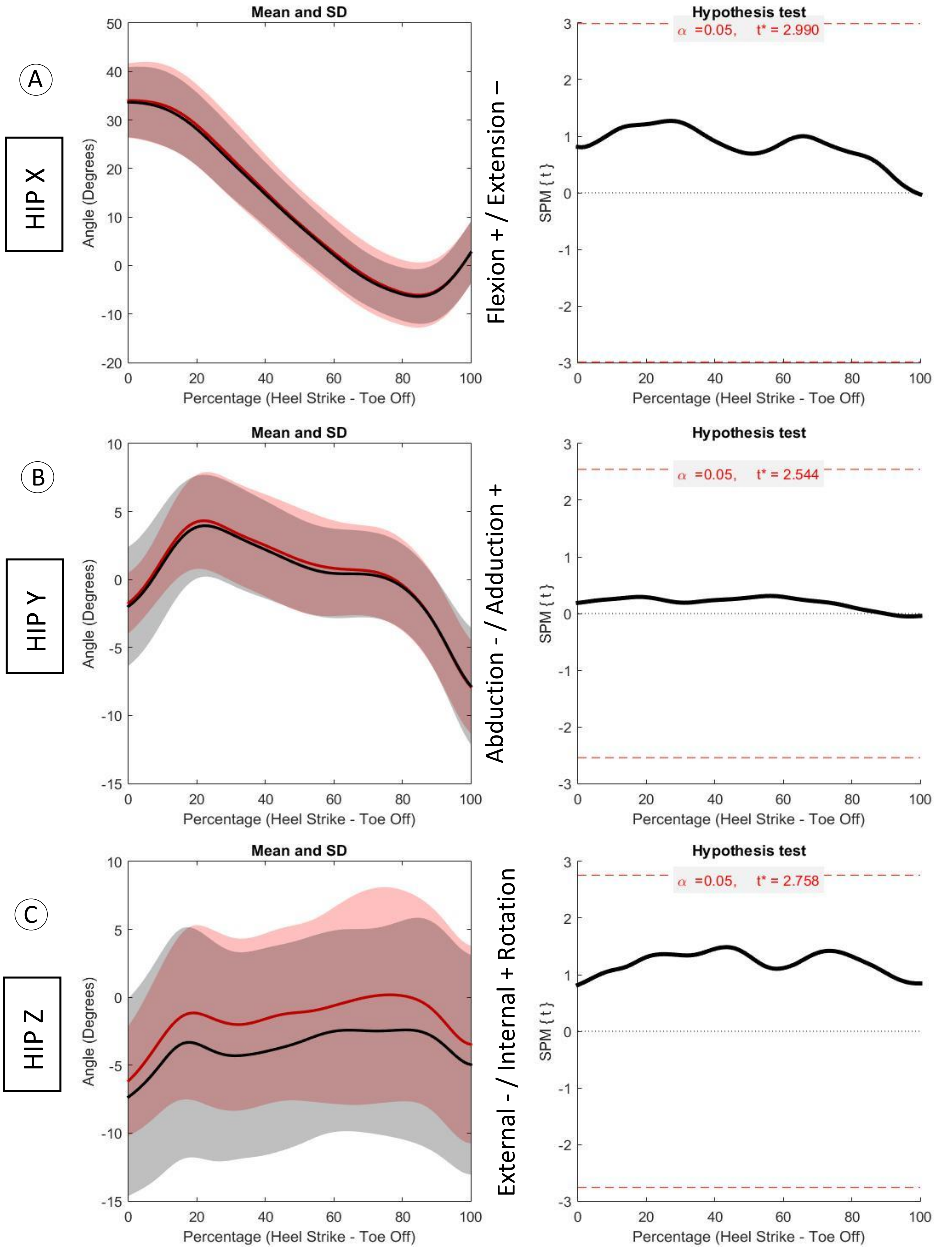


Figure E.17 Walking - Hip angles (x, y, z) Heel strike - Toe off – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

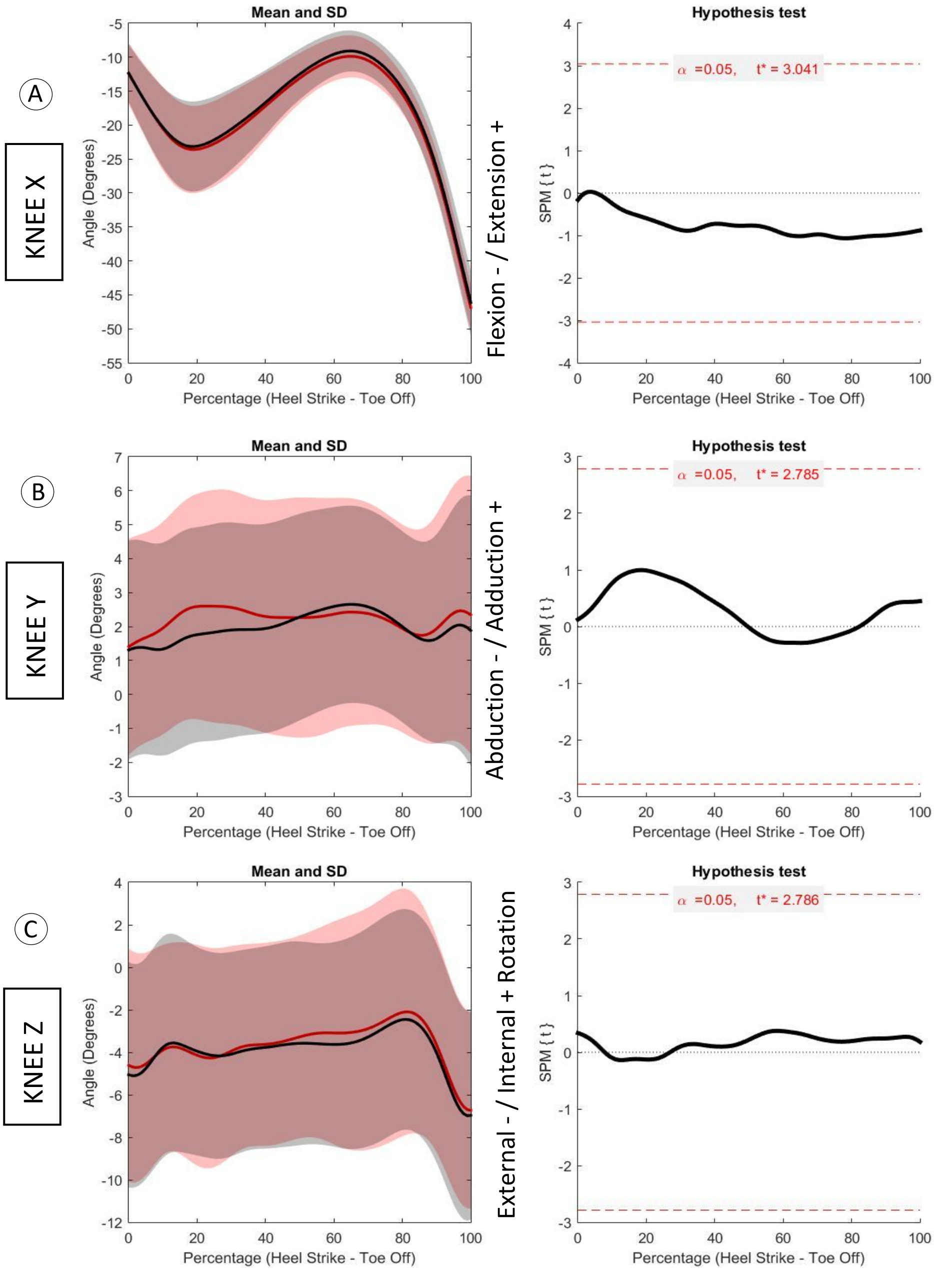


Figure E.18 Walking - Knee angles (x, y, z) Heel strike - Toe off – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

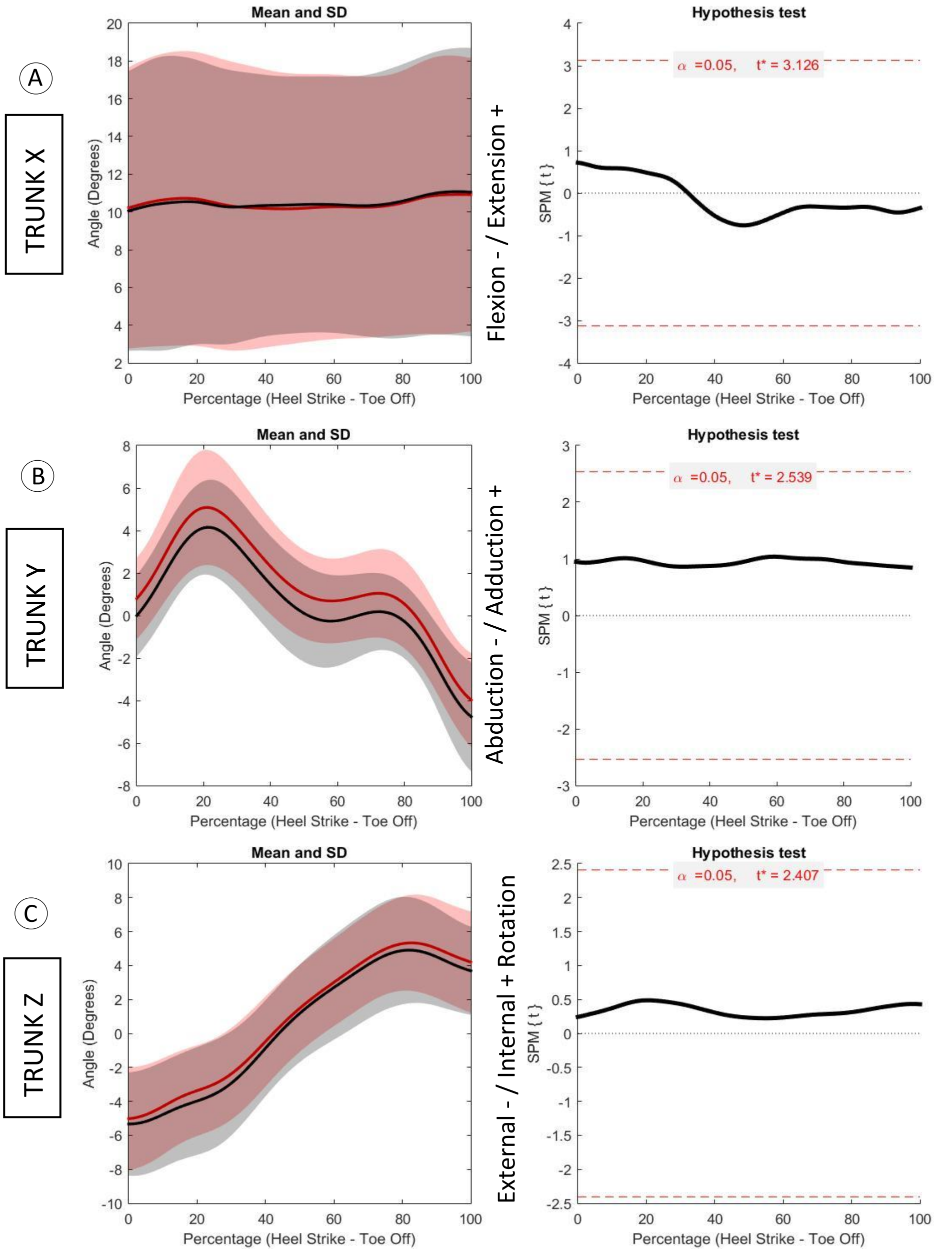


Figure E.19 Walking - Trunk angles (x, y, z) Heel strike - Toe off – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

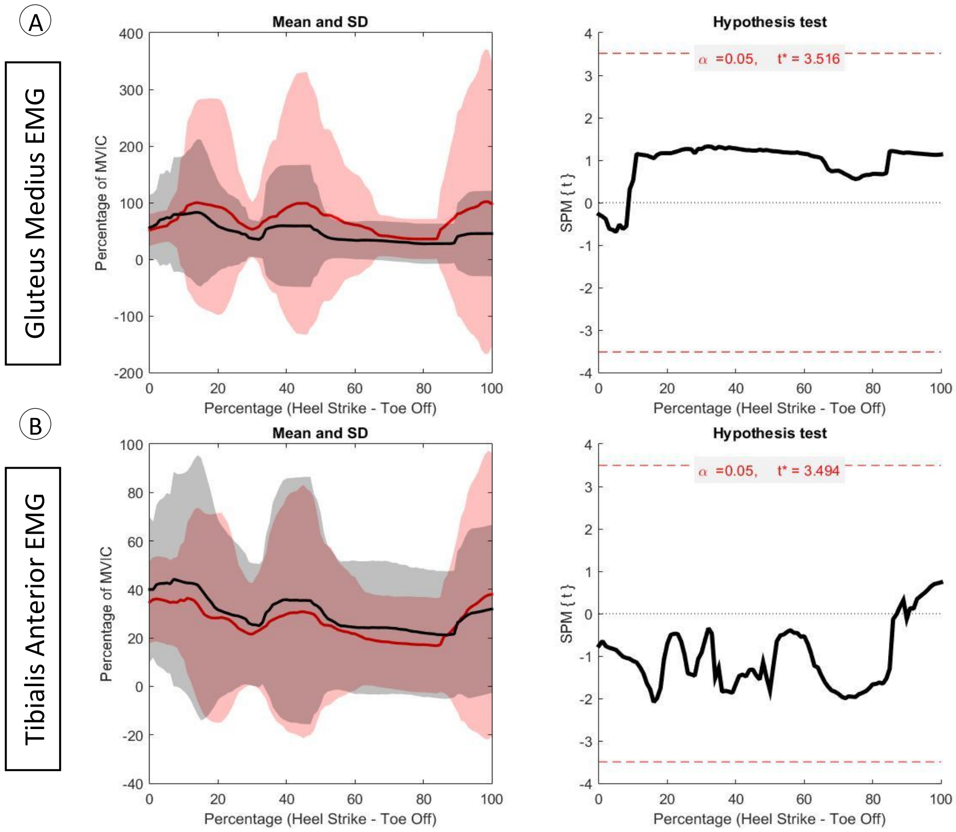


Figure E.20 Walking - Gluteus Medius and Tibialis Anterior muscle activation – heel strike - toe off means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

- **Toe off to Heel strike**

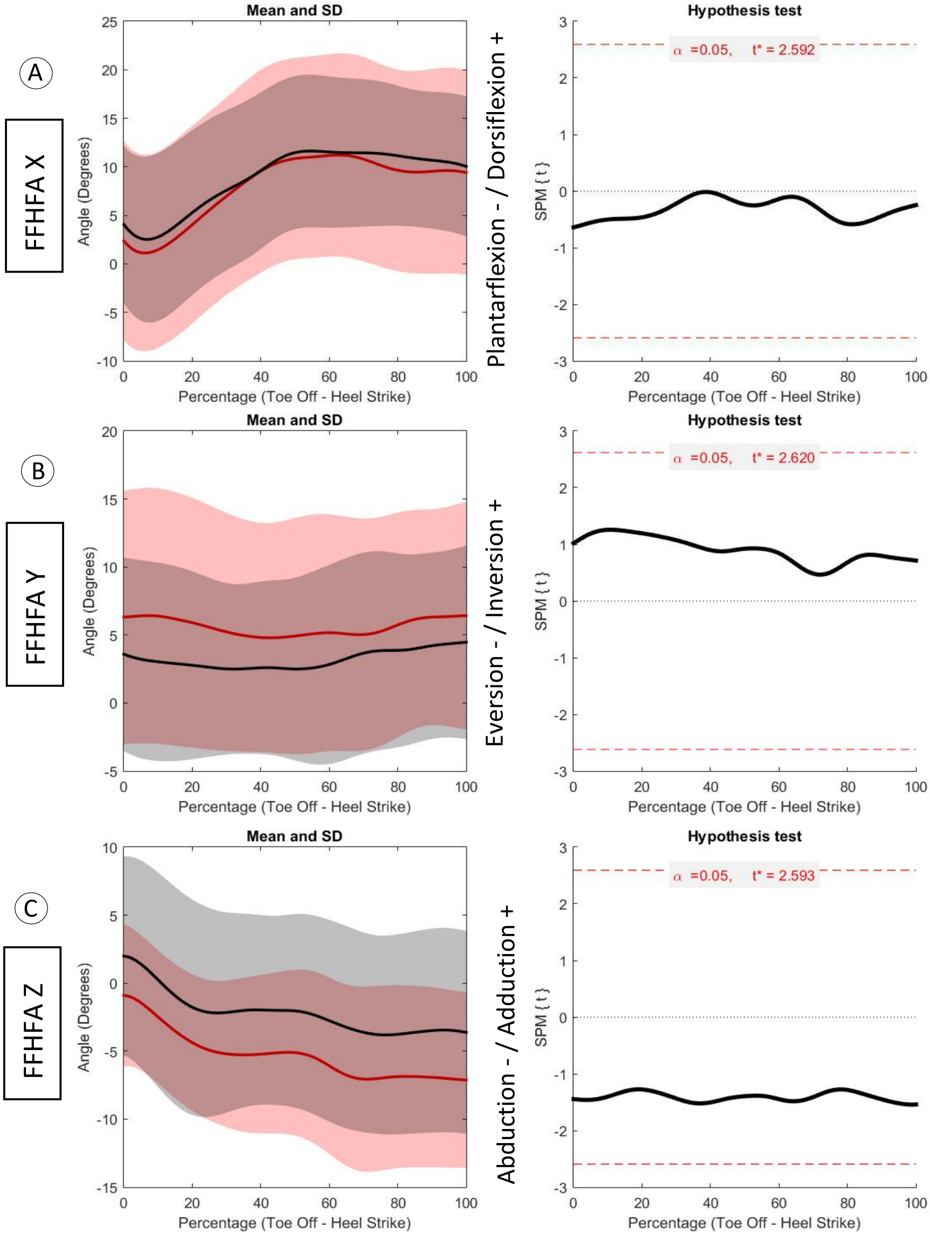


Figure E.21 Walking - Forefoot – Hindfoot angles (x, y, z) Toe off – Heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

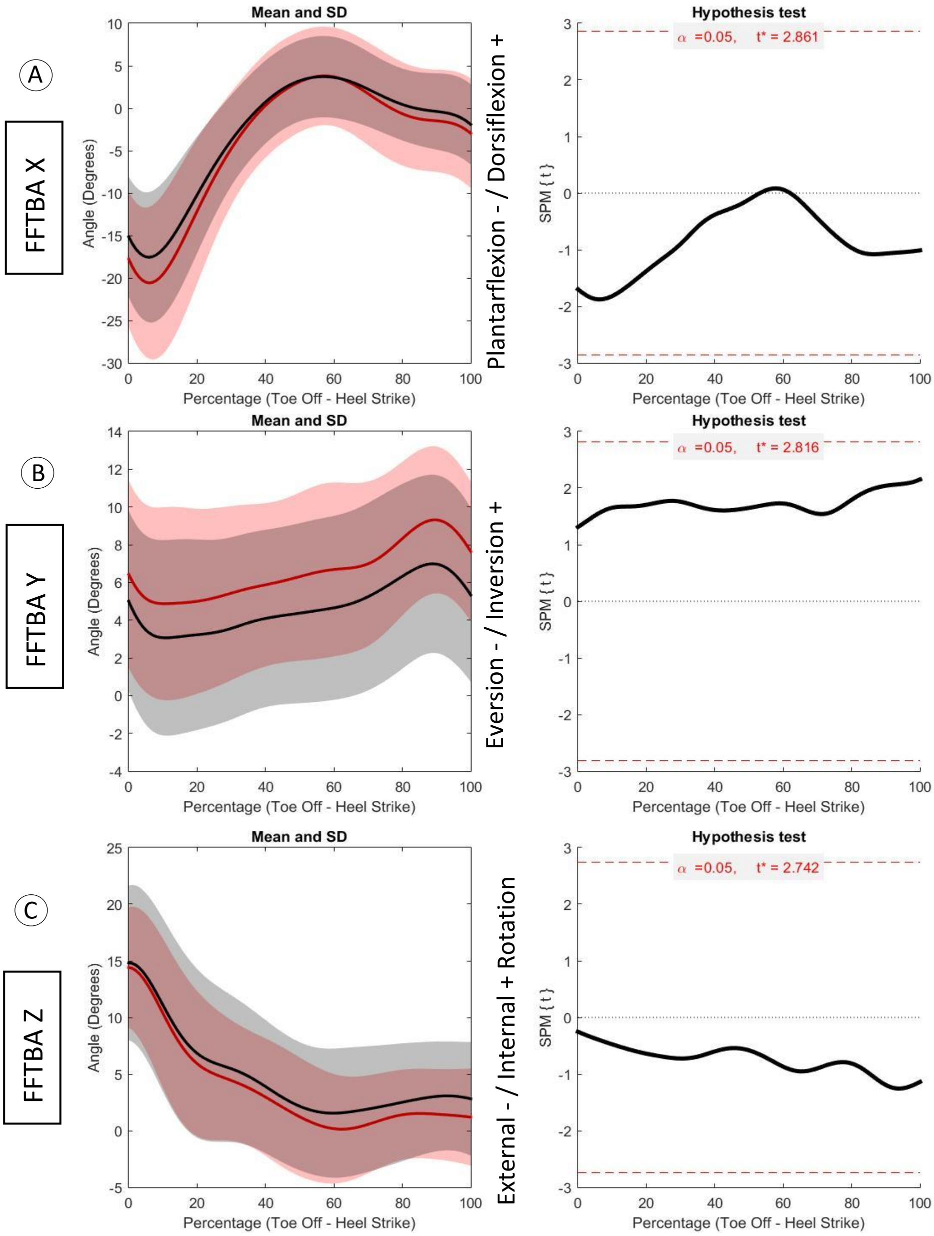


Figure E.22 Walking - Forefoot – Tibia angles (x, y, z) Toe off – Heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

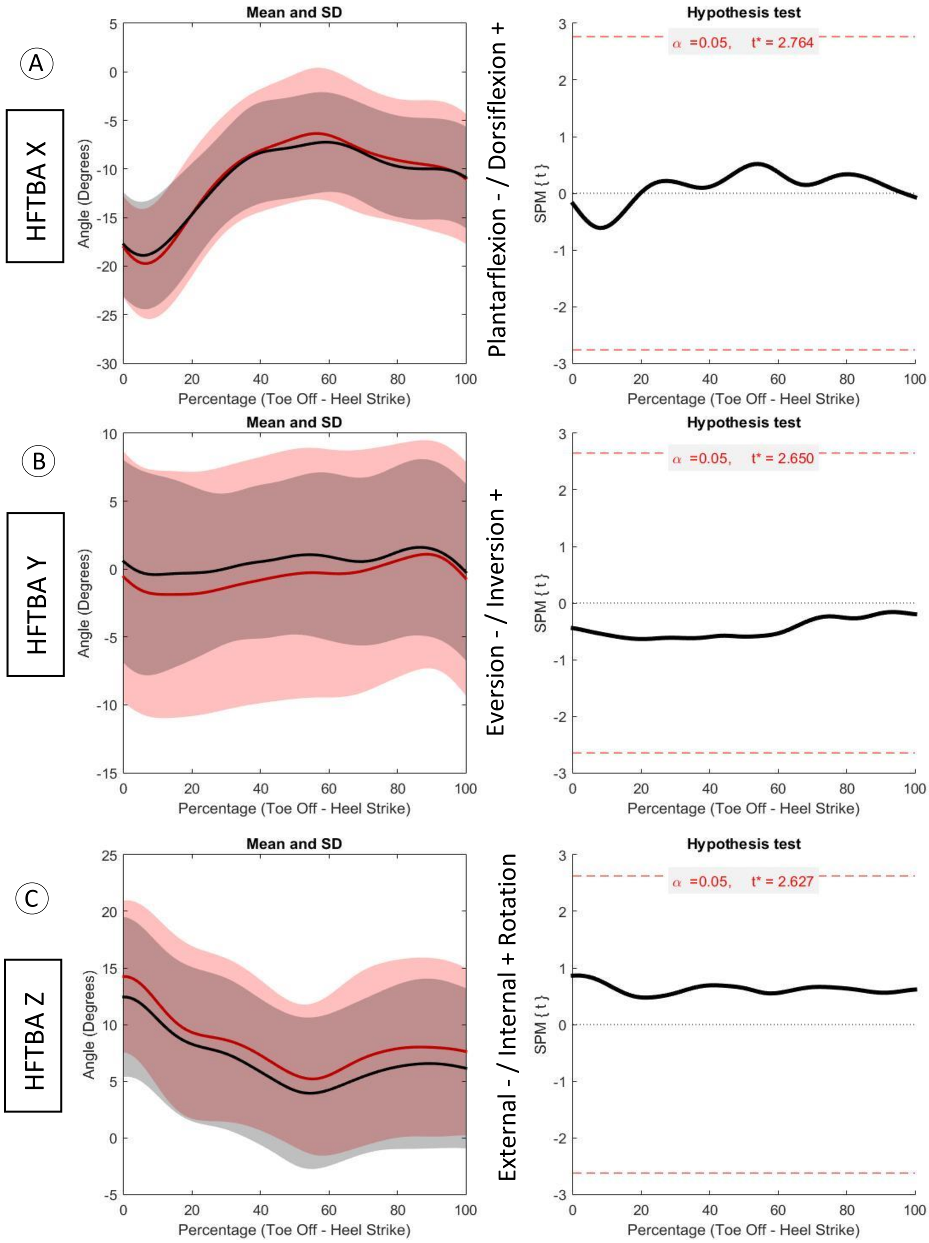


Figure E.23 Walking - Hindfoot – Tibia angles (x, y, z) Toe off – Heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

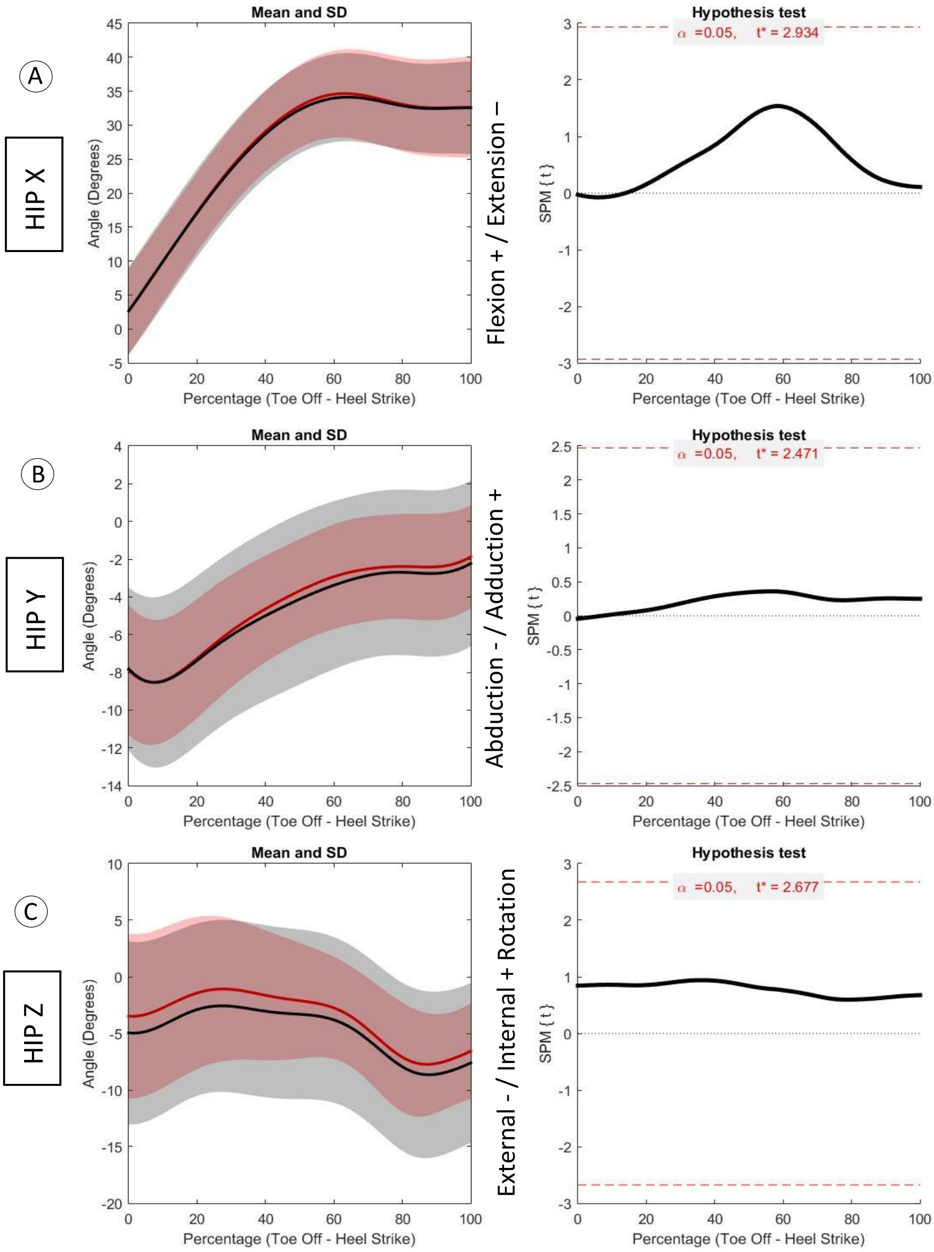


Figure E.24 Walking - Hip angles (x, y, z) Toe off – Heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

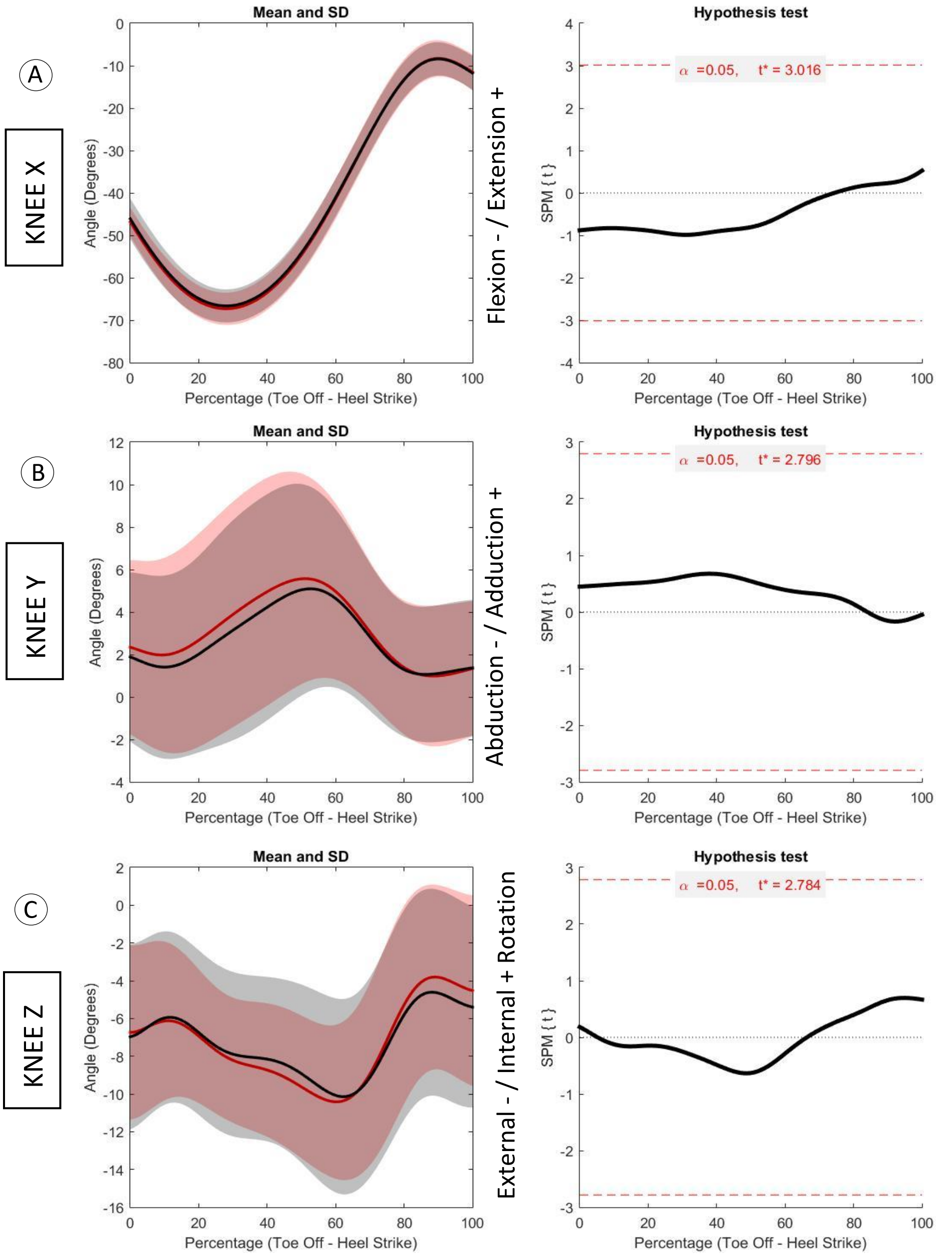


Figure E.25 Walking - Knee angles (x, y, z) Toe off – Heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

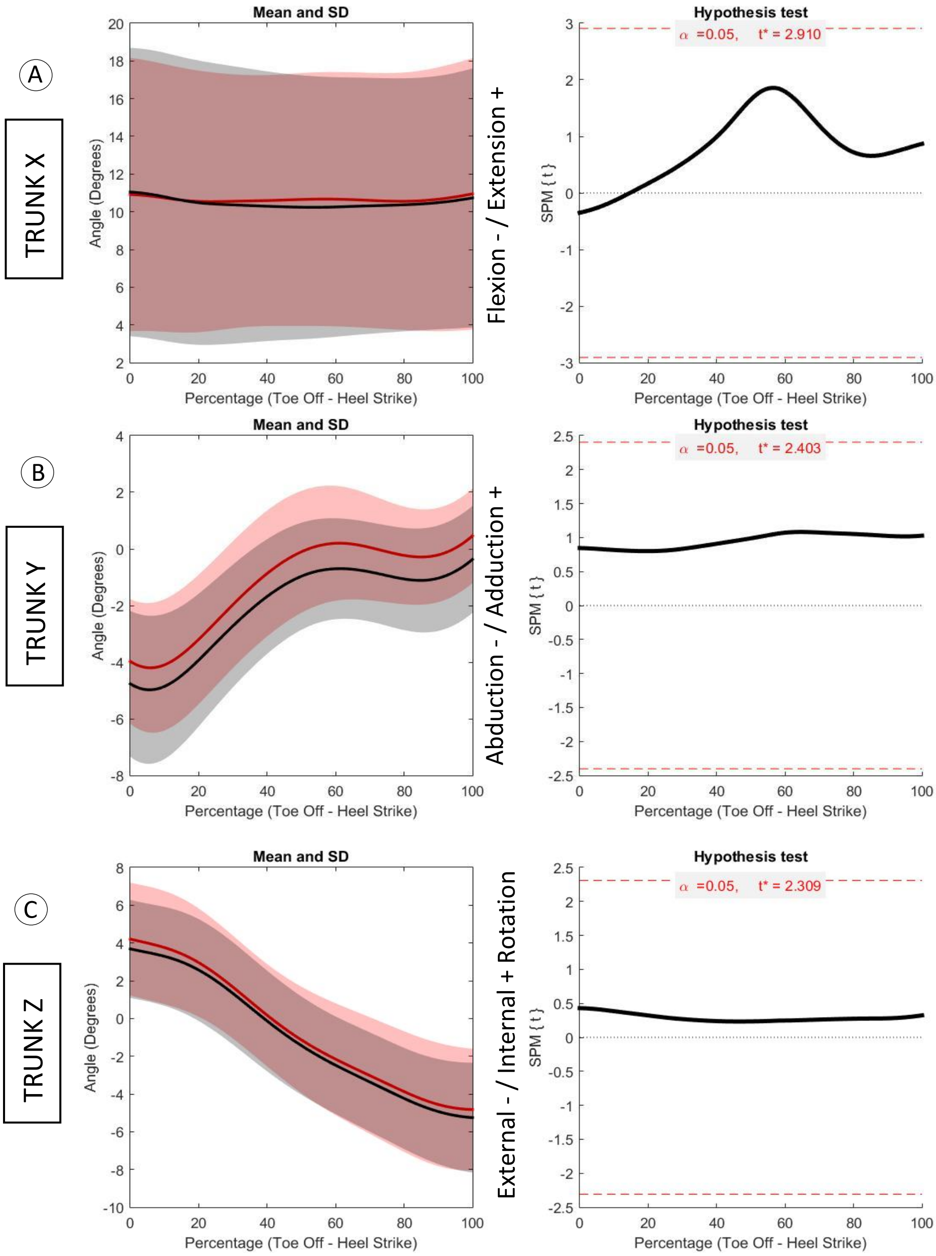


Figure E.26 Walking - Knee angles (x, y, z) Toe off – Heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

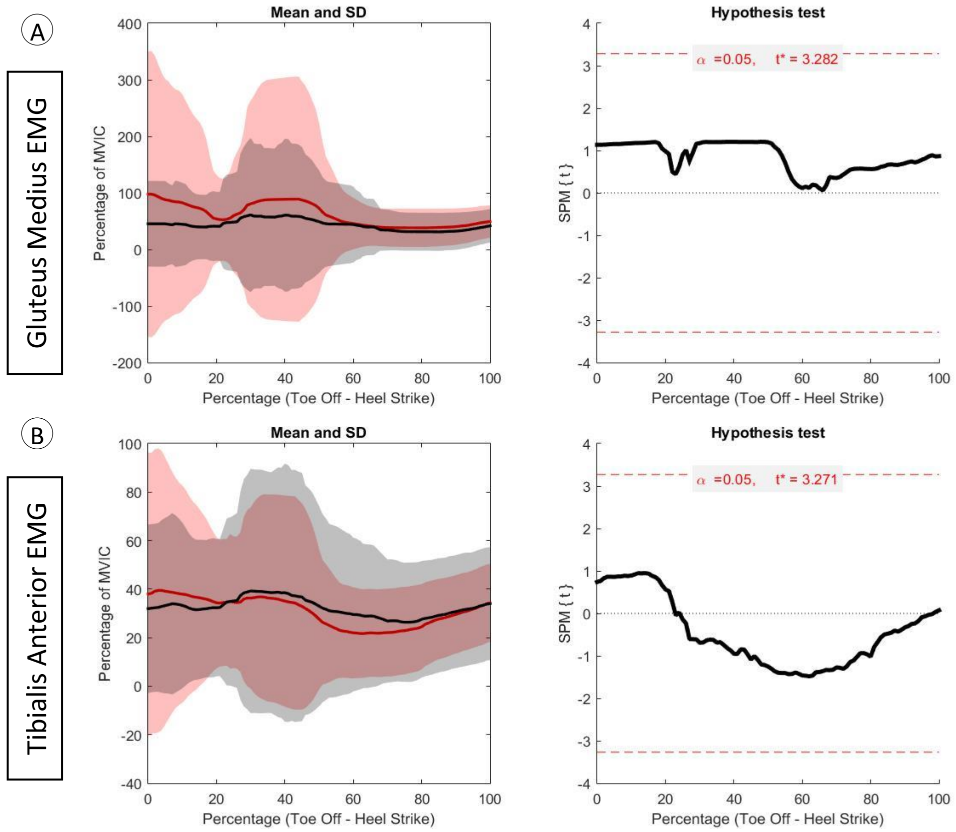


Figure E.27 Walking - Gluteus Medius and Tibialis Anterior muscle activation –toe off - heel strike - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

Matched Control vs Unaffected limb of CAI group

- **Heel strike to Toe off**

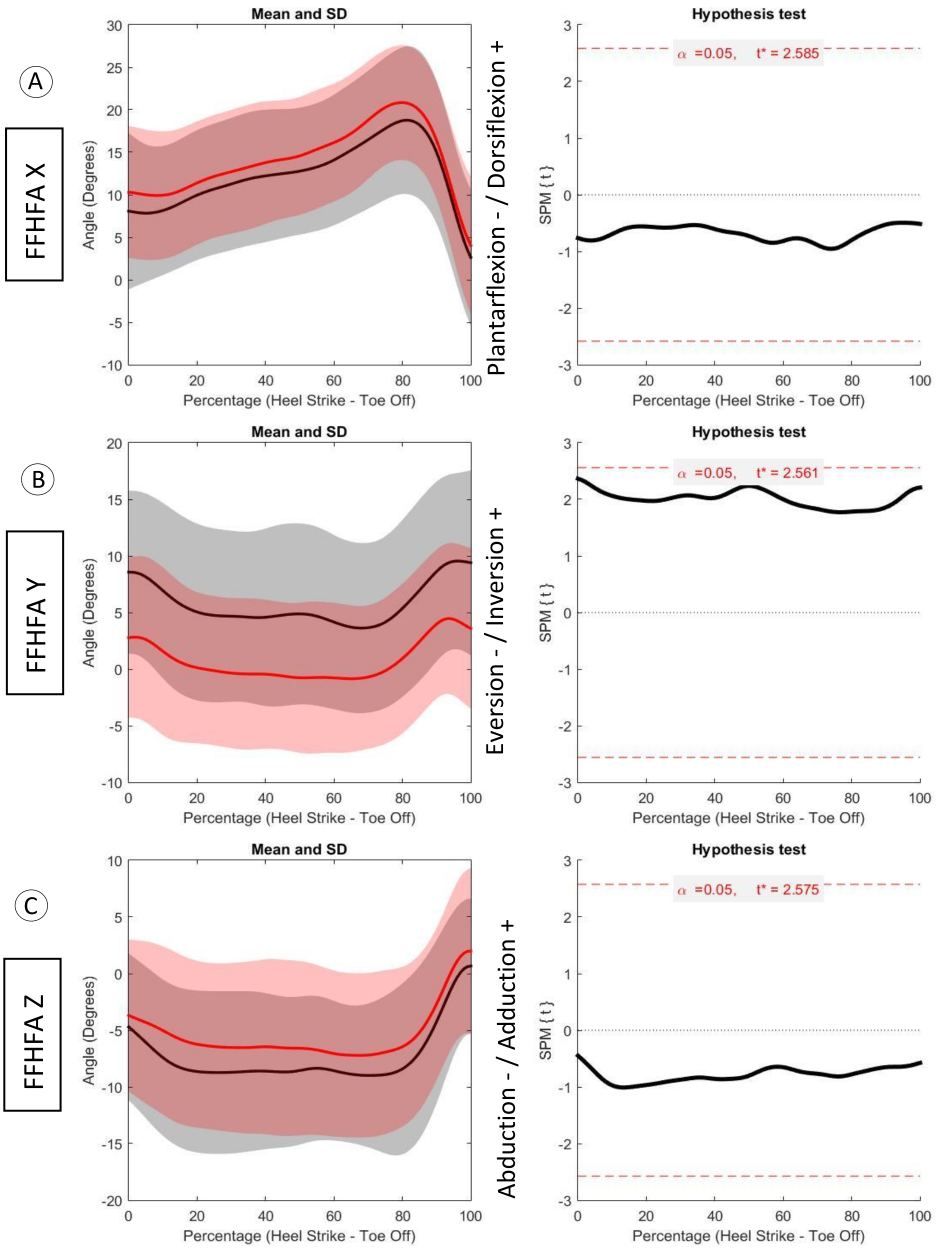


Figure E.28 Walking - Forefoot – Hindfoot angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

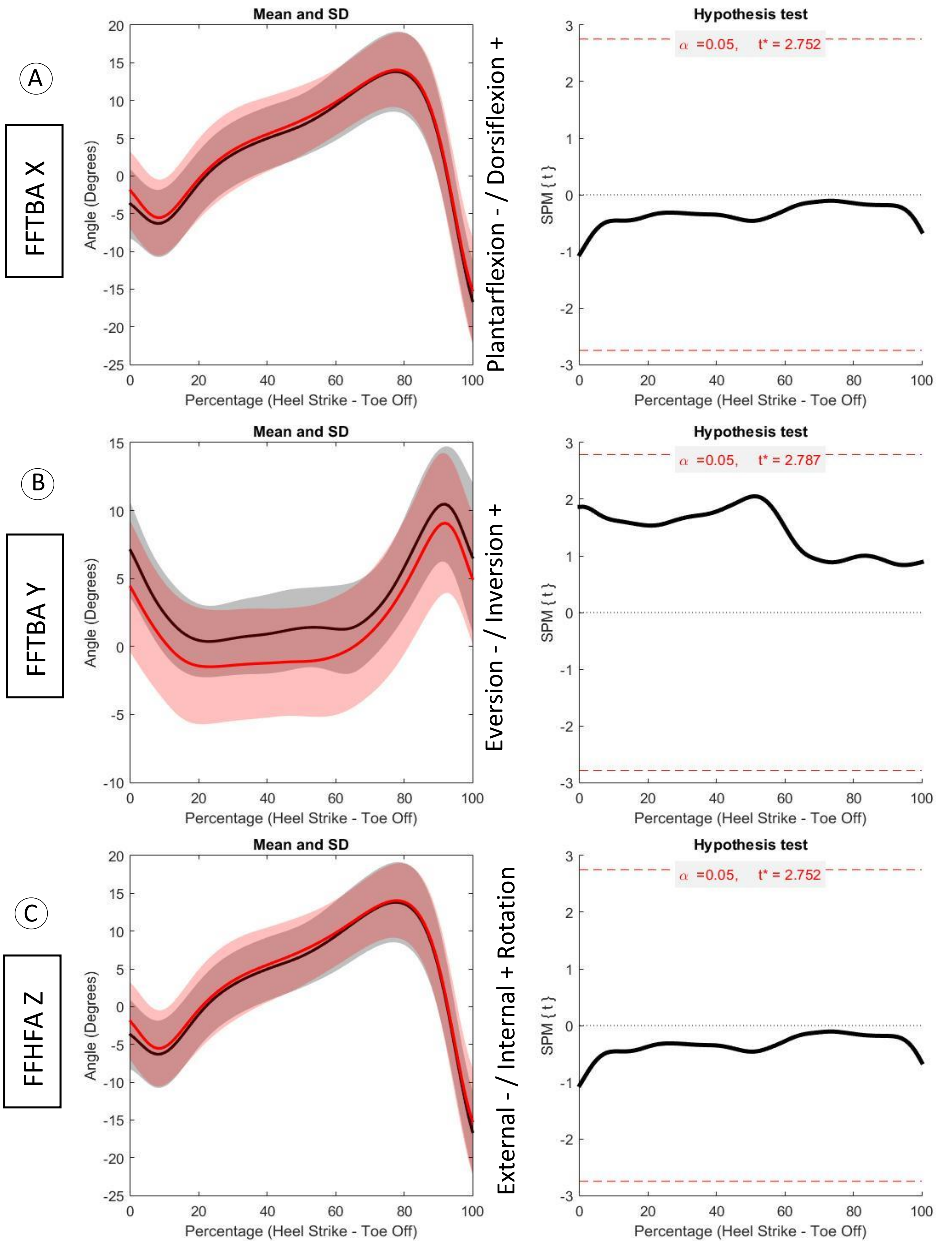


Figure E.29 Walking - Forefoot – tibia angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

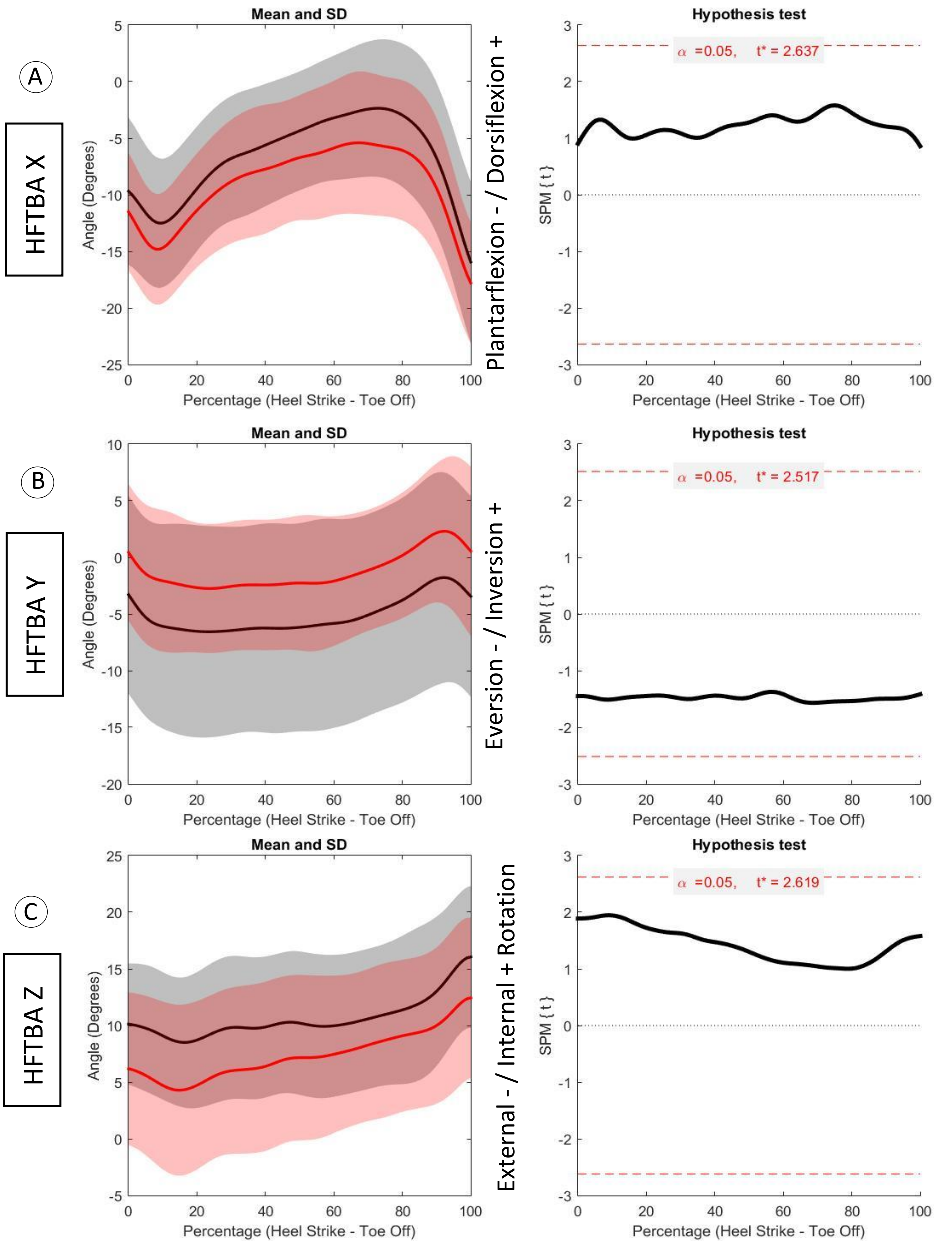


Figure E.30 Walking - Hindfoot – tibia angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

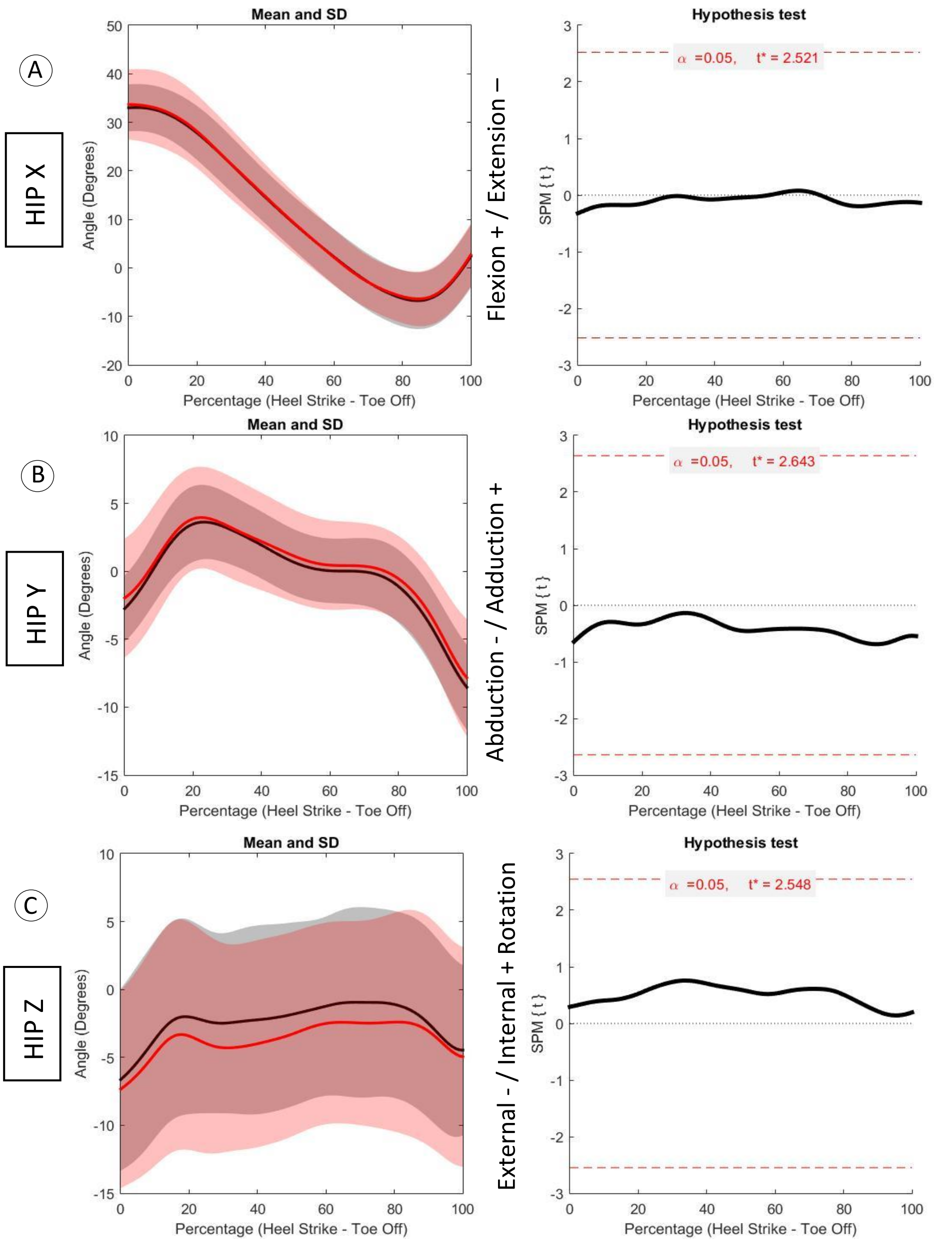


Figure E.31 Walking - Hip angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

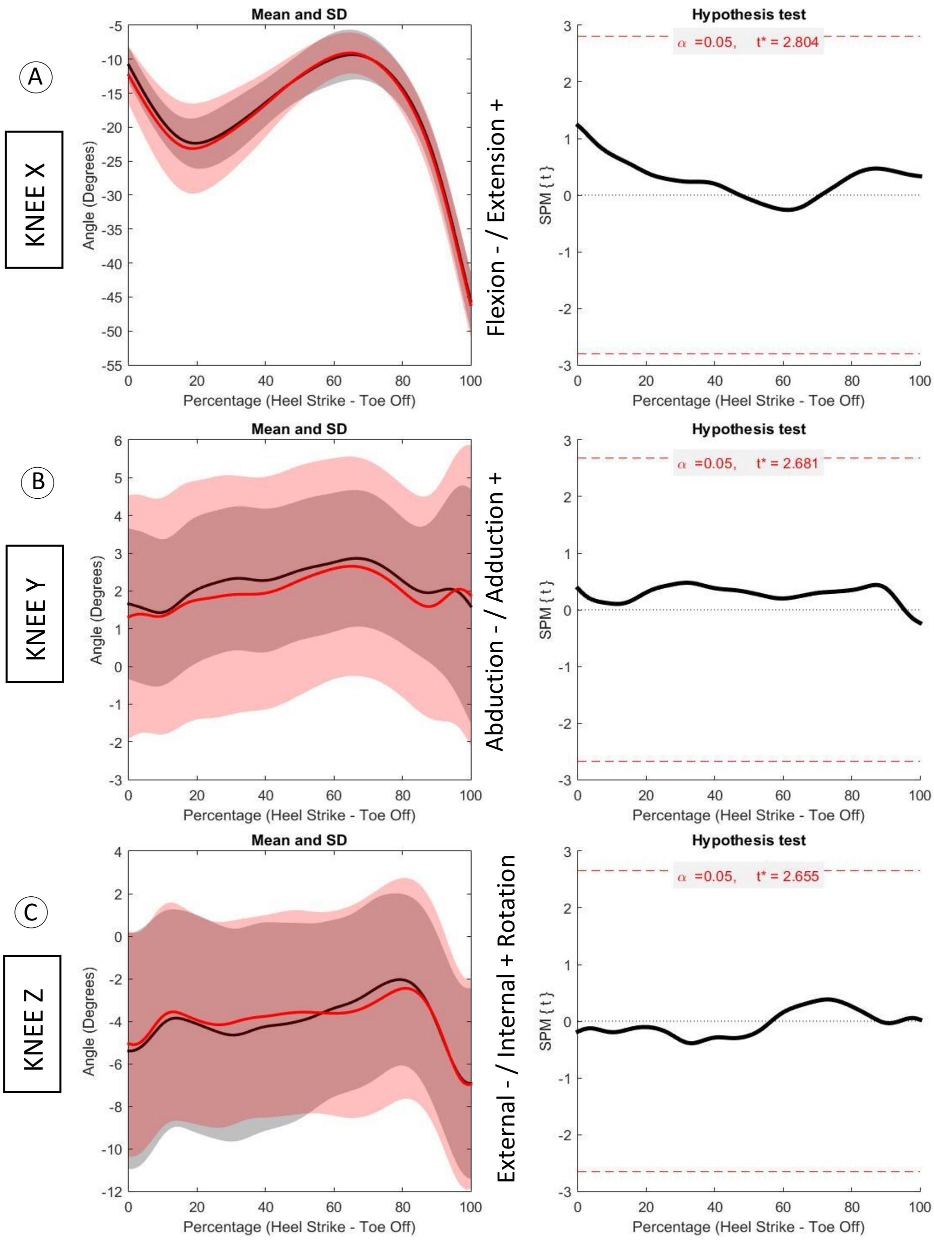


Figure E.32 Walking - Knee angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

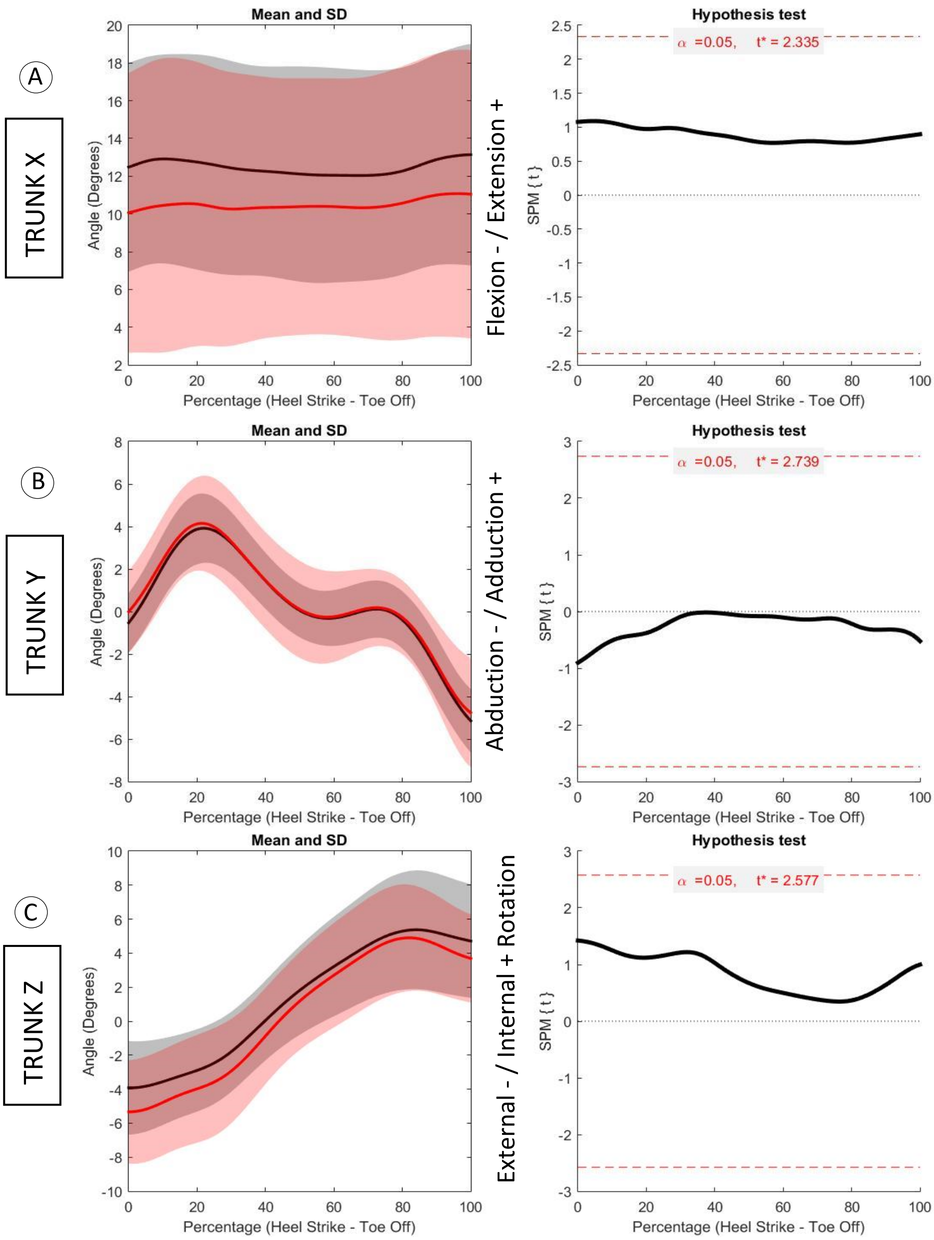


Figure E.33 Walking - Trunk angles (x, y, z) Heel strike – toe off - means and standard deviations (Matched Control - — Unaffected - —) and t-test output.

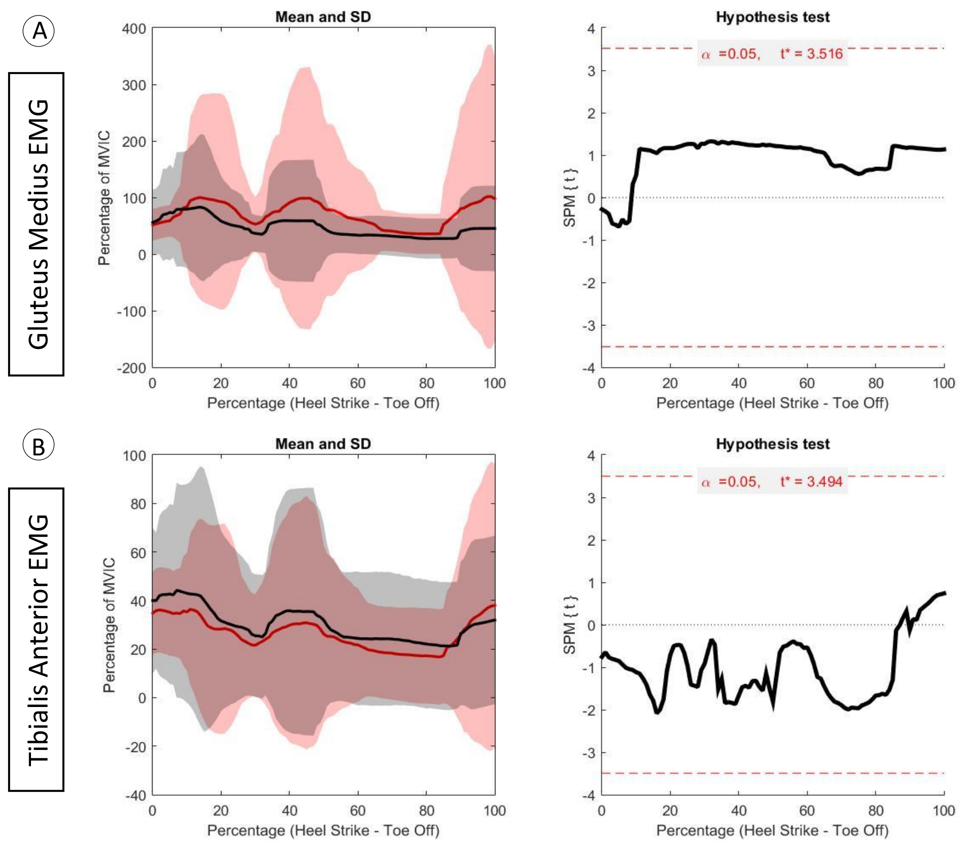


Figure E.34 Walking - Gluteus Medius and Tibialis Anterior muscle activation – heel strike - toe off means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

- **Toe off to Heel strike**

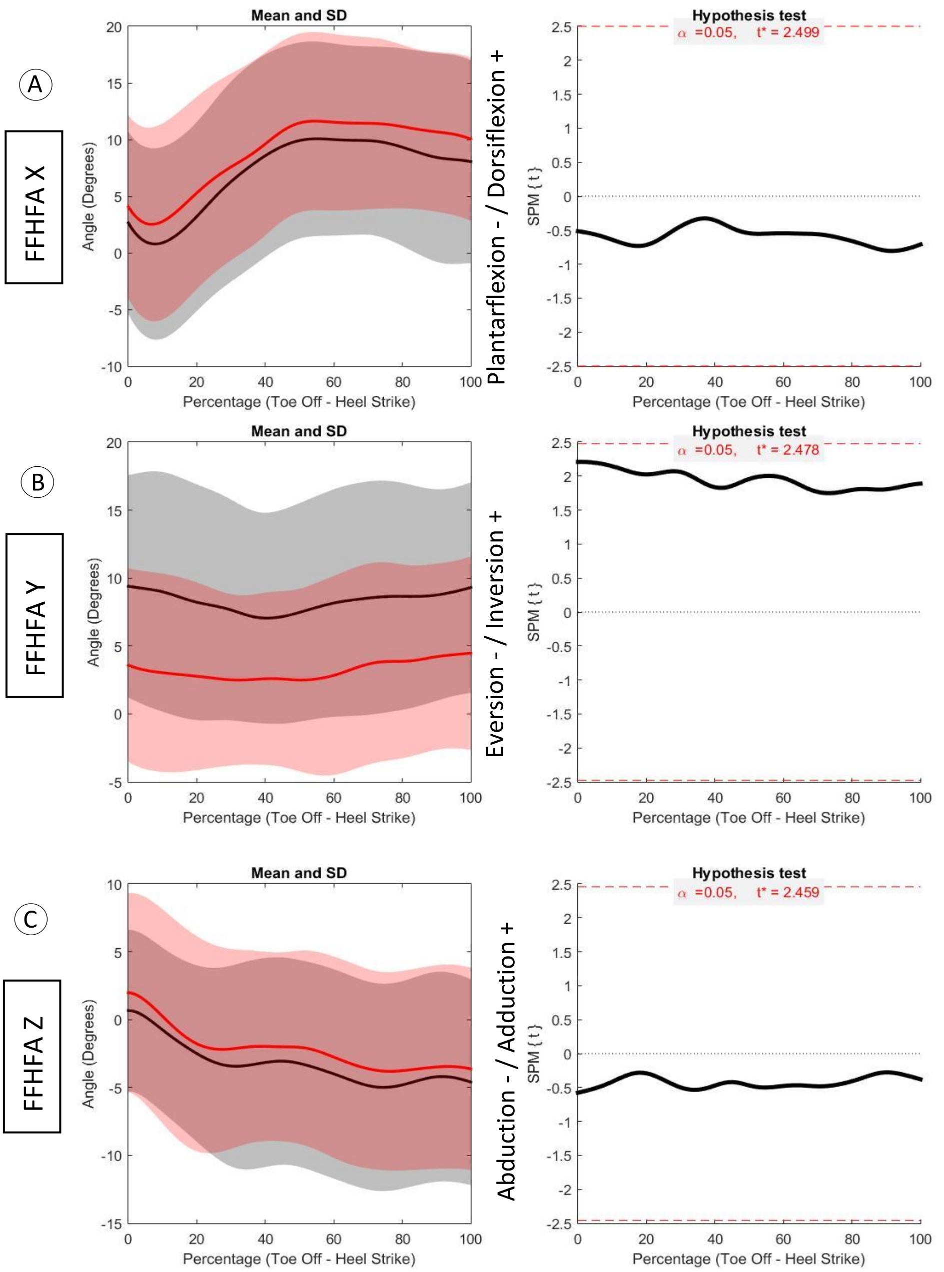


Figure E.35 Walking - Forefoot – Hindfoot angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

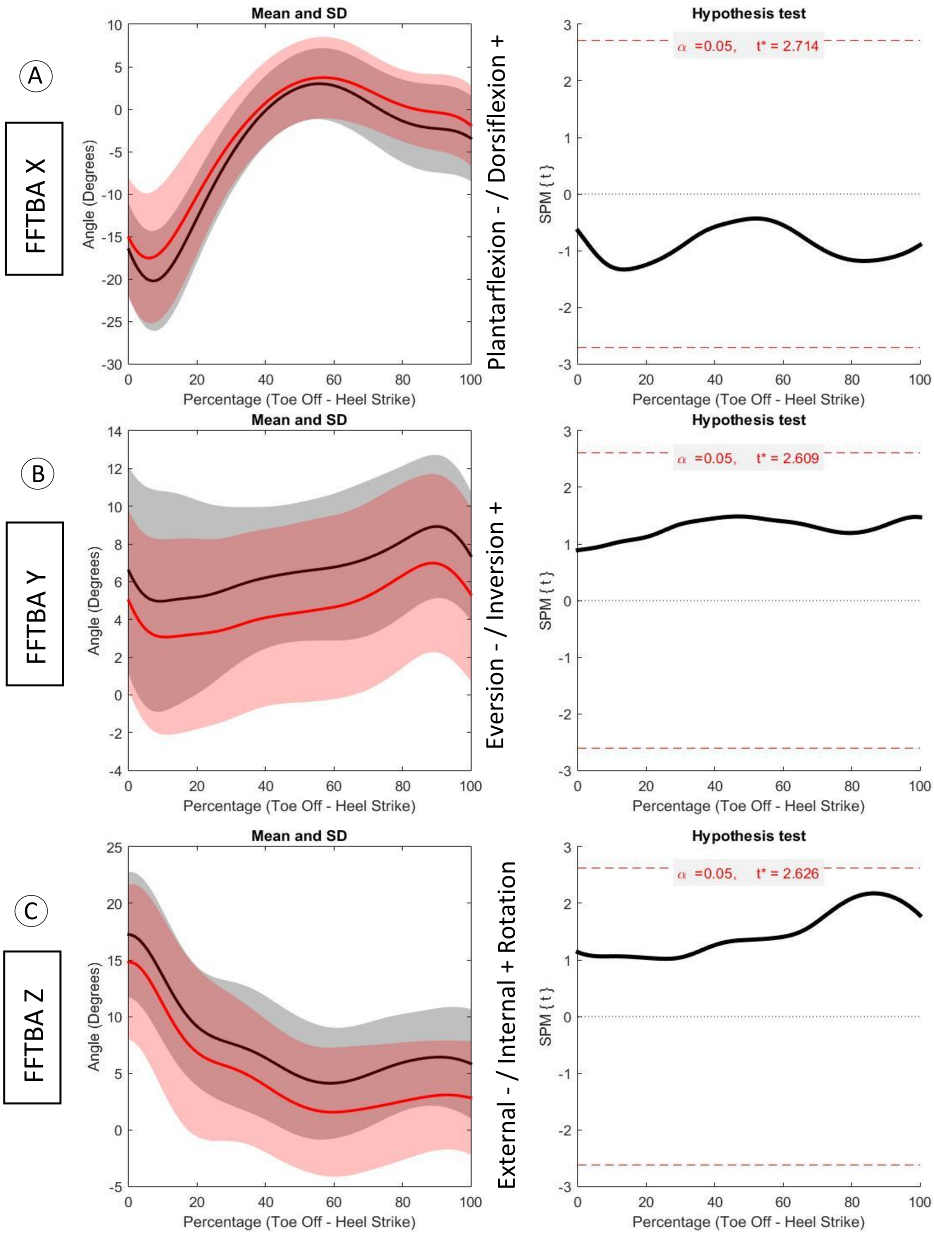


Figure E.36 Walking - Forefoot –Tibia angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

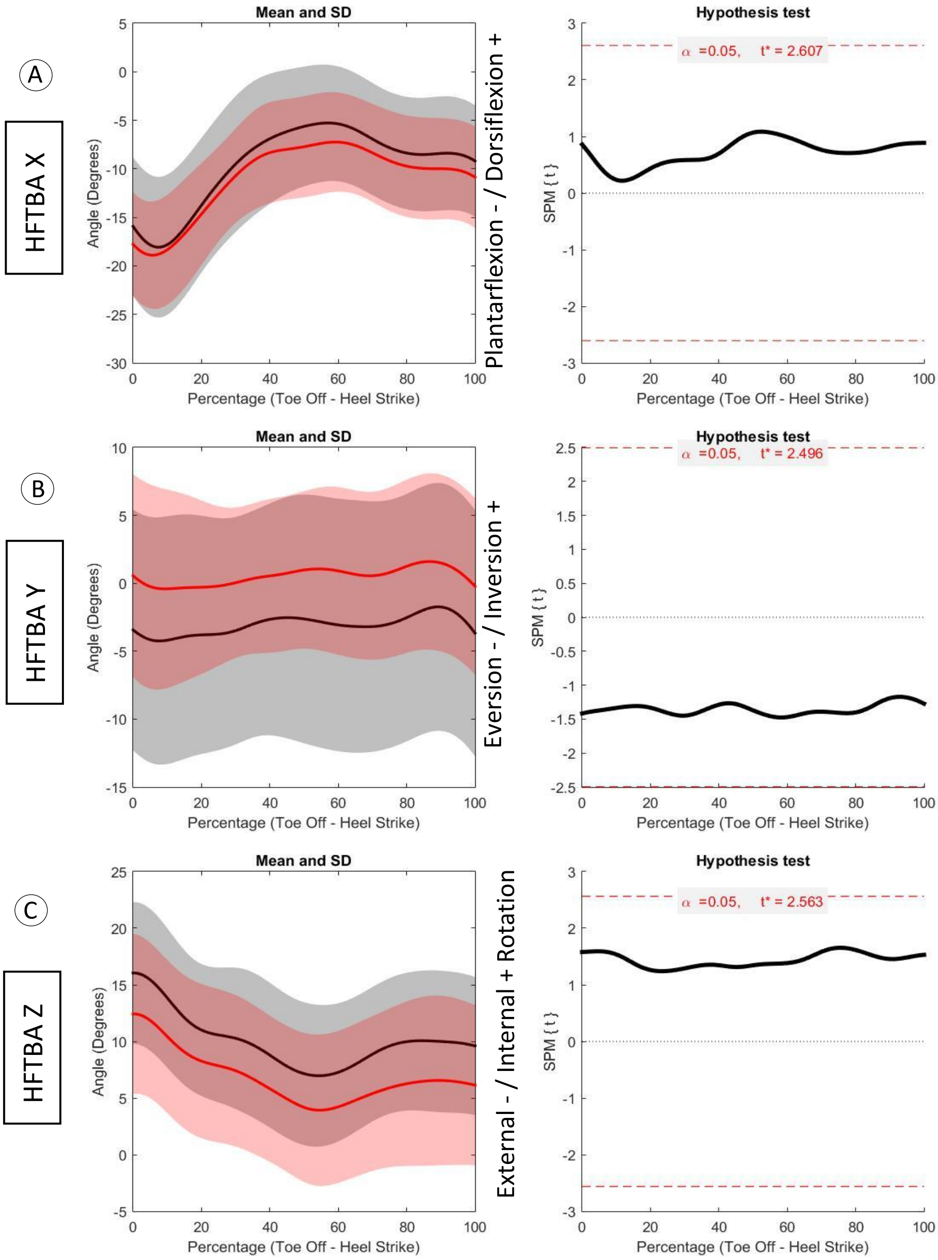


Figure E.37 Walking - Hindfoot-Tibia angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

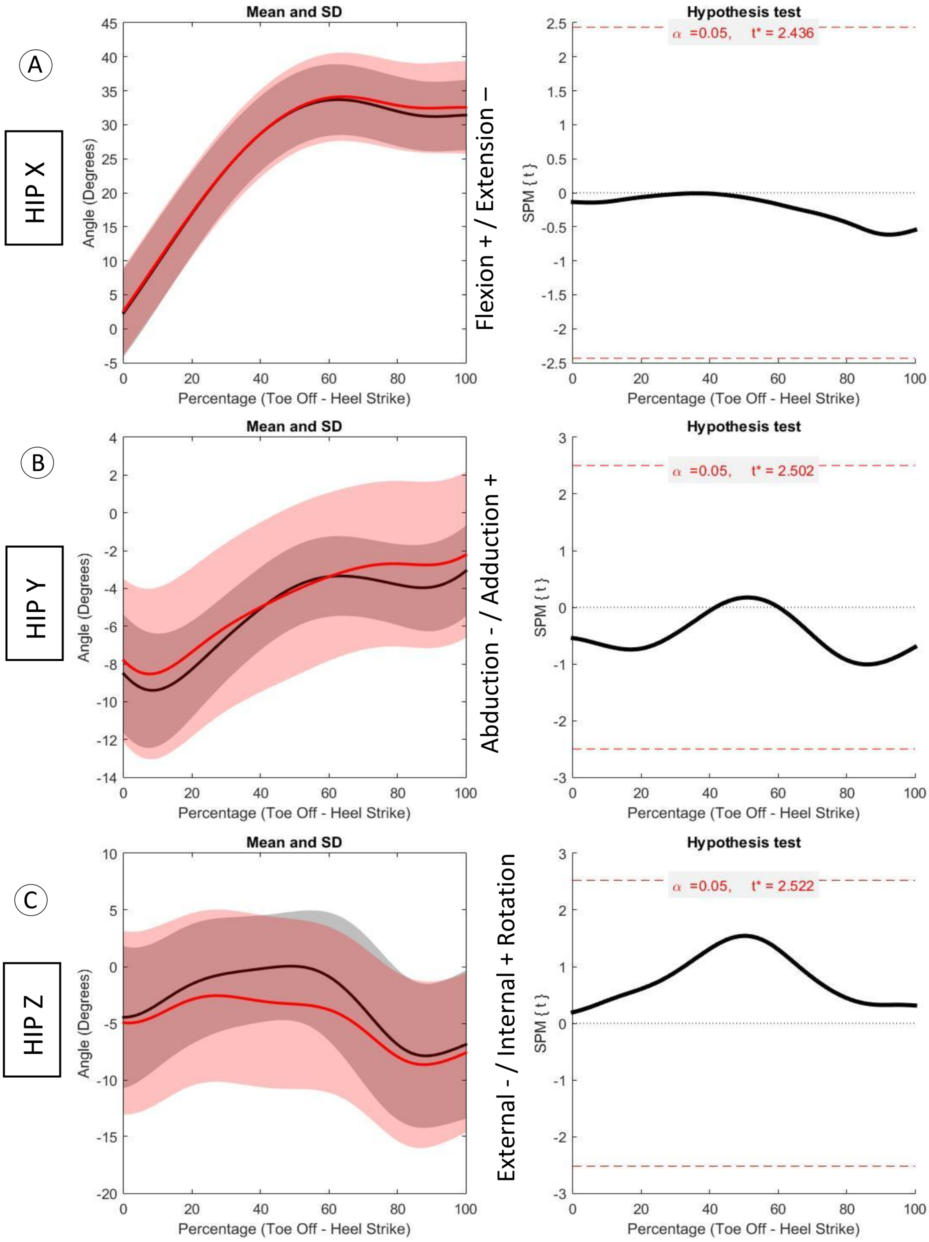


Figure E.38 Walking – Hip angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

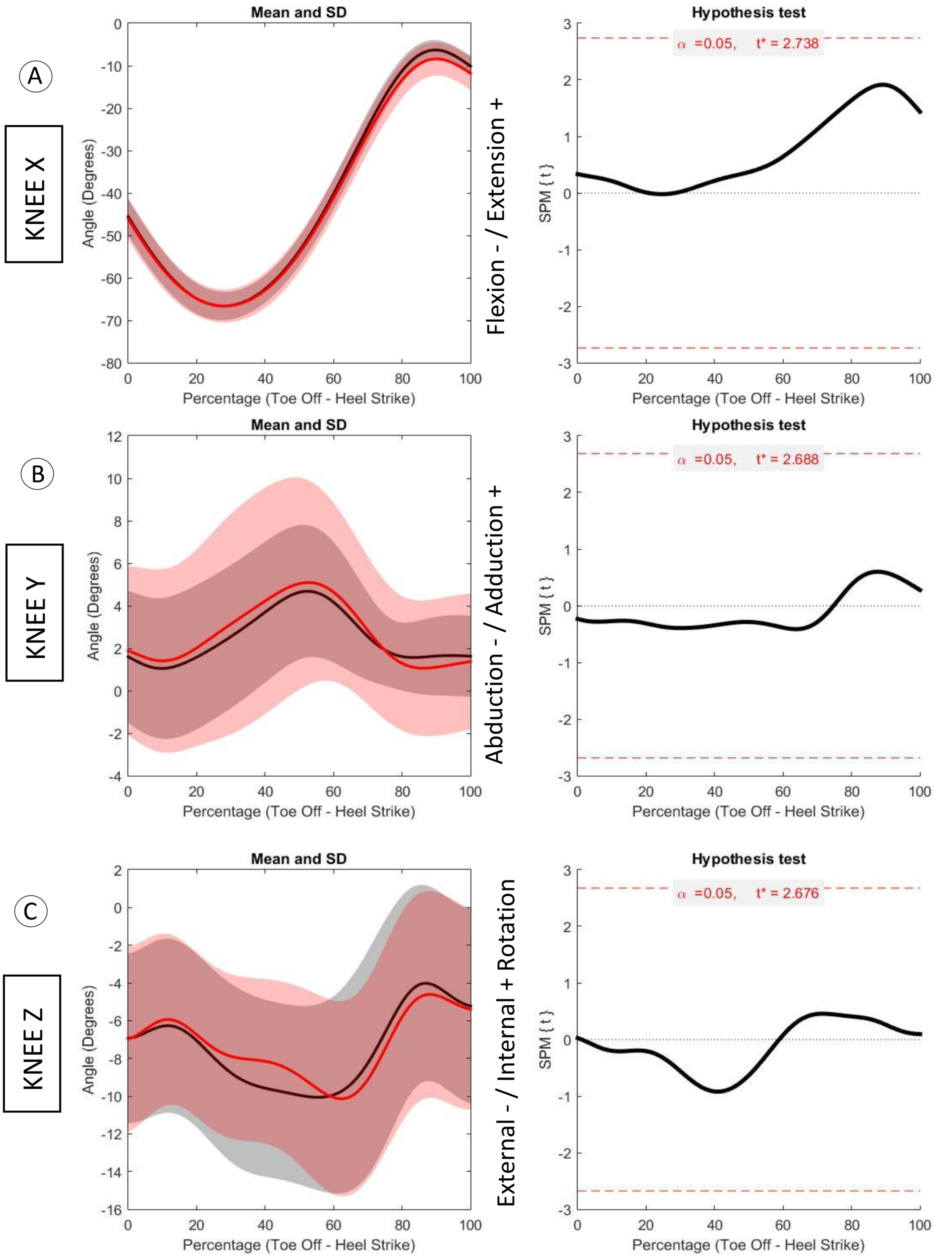


Figure E.39 Walking - Knee angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

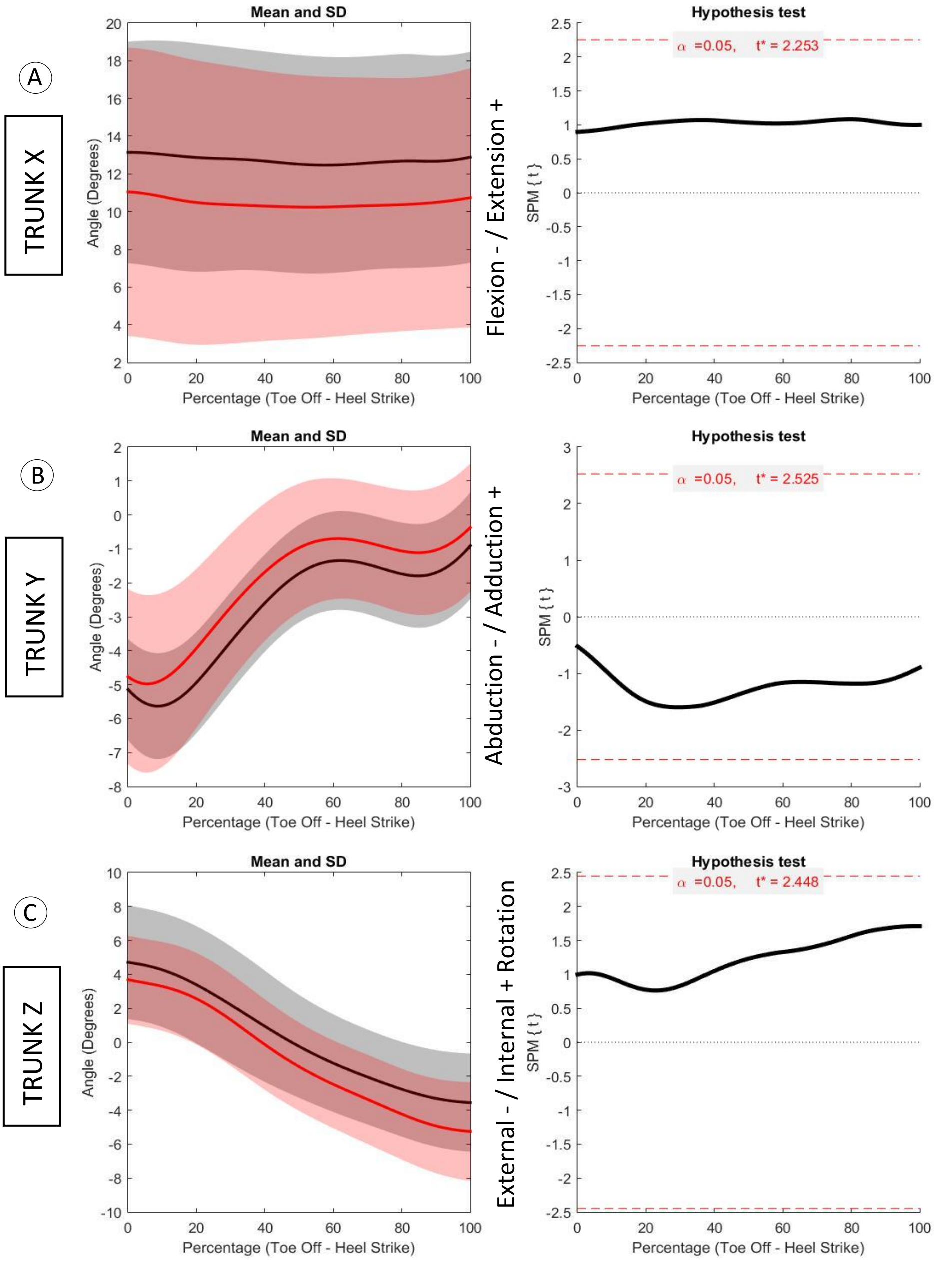


Figure E.40 Walking - Trunk angles (x, y, z) Toe off – Heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

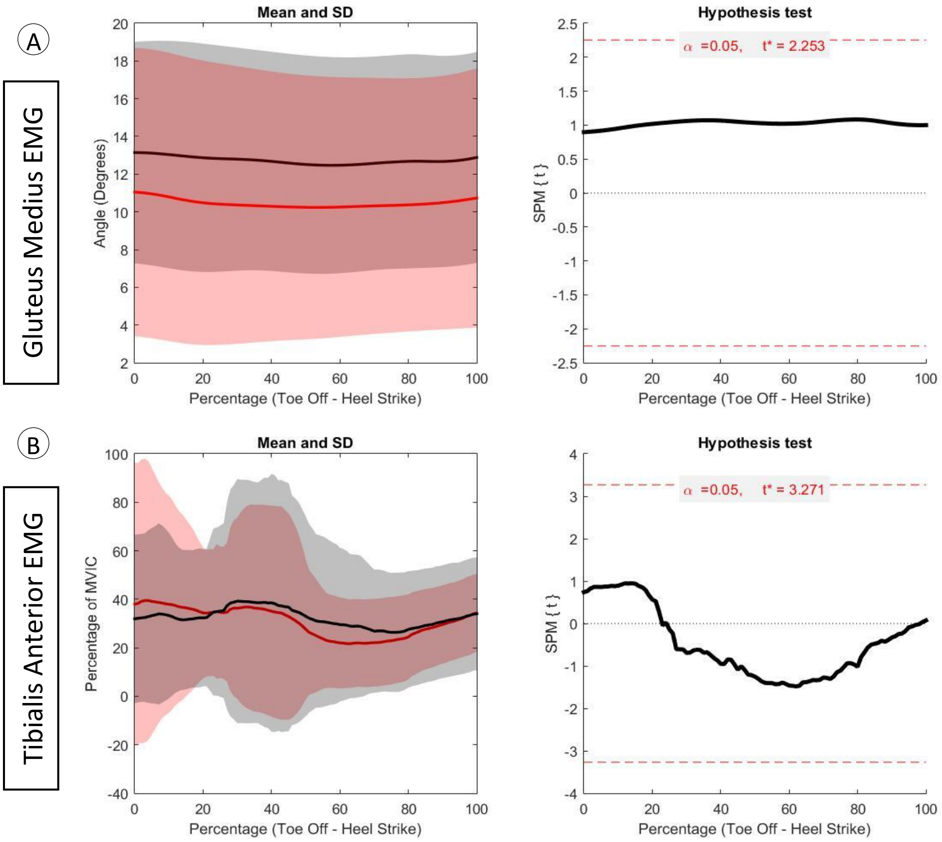


Figure E.41 Walking - Gluteus Medius and Tibialis Anterior muscle activation – toe off - heel strike - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

Appendix F Study 1 Part 2 – Walk Discrete

Angular Displacement

Table F.1 Peak angular displacement from 100 ms pre-initial contact to heel strike (degrees)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	9.66	10.10	8.00	9.30	11.03	11.53	12.06	7.49
	Plantarflexion	7.12	9.56	5.65	9.22	8.05	10.68	9.39	7.75
FFHFA Frontal	Inversion	9.28	7.40	7.35	9.04	6.55	9.06	4.42	7.27
	Eversion	7.21	7.52	4.94	9.12	4.44	8.84	2.11	7.38
FFHFA Transverse	Adduction	-3.68	7.20	-3.57	7.74	-6.08	7.01	-2.94	7.29
	Abduction	-5.58	7.58	-5.47	7.55	-8.15	7.31	-4.86	7.23
FFTBA Sagittal	Dorsiflexion	-0.89	5.11	-0.43	4.97	-0.13	6.24	0.88	4.79
	Plantarflexion	-3.88	5.14	-3.40	5.41	-3.05	6.48	-2.02	5.22
FFTBA Frontal	Inversion	9.08	3.97	9.20	4.99	9.18	4.19	6.69	4.78
	Eversion	6.66	3.91	6.79	5.11	6.74	3.84	4.33	4.95
FFTBA Transverse	Internal Rotation	7.25*	4.37	3.95	4.15	2.30*	4.33	3.70	5.40
	External Rotation	5.37*	4.22	1.79	4.03	0.60*	4.18	1.83	5.22
HFTBA Sagittal	Dorsiflexion	-7.45	7.19	-6.45	6.35	-8.28	6.86	-9.03	5.44
	Plantarflexion	-10.13	7.39	-9.04	6.61	-11.46	6.84	-11.76	5.48
HFTBA Frontal	Inversion	-1.06	9.00	2.08	7.92	1.78	9.46	2.35	6.37
	Eversion	-3.63	9.02	-0.67	8.42	-0.86	9.46	-0.55	6.63
HFTBA Transverse	Internal Rotation	11.45	5.91	7.94	6.30	8.92	8.30	7.66	7.29
	External Rotation	9.33	5.83	5.81	6.56	6.94	7.82	5.56	7.26
HIP Sagittal	Extension	33.67	5.30	33.66	5.25	34.63	7.20	34.58	6.76
	Flexion	32.05	5.16	31.82	5.08	32.50	7.67	32.61	7.14
HIP Frontal	Adduction	-2.57	2.49	-1.25	2.89	-1.55	2.52	-1.93	4.41
	Abduction	-3.88	2.41	-2.58	2.96	-2.58	2.63	-2.96	4.57
HIP Transverse	Internal Rotation	-5.36	6.79	-6.26	7.37	-5.49	4.41	-6.78	7.75
	External Rotation	-8.42	6.83	-9.29	7.54	-8.32	4.27	-9.08	7.62
KNEE Sagittal	Extension	-6.71	2.58	-7.01	3.02	-8.38	4.57	-8.27	4.31
	Flexion	-14.33	3.81	-15.12	3.79	-16.25	4.93	-15.92	3.80
KNEE Frontal	Adduction	2.20	1.79	2.38	2.13	2.01	3.16	1.82	3.23
	Abduction	1.28	1.82	1.46	2.01	0.80	3.36	0.79	3.38
KNEE Transverse	Internal Rotation	-4.12	5.60	-4.14	6.39	-3.61	5.40	-3.90	5.38
	External Rotation	-6.59	5.84	-6.53	6.28	-6.44	5.31	-6.69	5.49
TRUNK Sagittal	Extension	13.13	5.64	12.65	5.68	10.89	7.39	10.81	7.21
	Flexion	12.29	5.60	11.83	5.68	10.04	7.24	10.03	7.13
TRUNK Frontal	Adduction	-0.77	1.70	0.66 †	1.33	0.44	1.97	-0.51 †	1.87
	Abduction	-1.62*	1.64	-0.01	1.47	0.05*	1.87	-0.80	2.26
TRUNK Transverse	Internal Rotation	-2.86	2.82	-4.43	3.34	-4.18	3.56	-4.74	2.79
	External Rotation	-3.67	2.98	-5.31	3.69	-4.49	3.05	-4.92	3.03

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Table F.2 Angular displacement at heel strike (degrees)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	6.63	9.29	8.01	9.51	9.30	10.91	10.33	7.73
FFHFA Frontal	6.57	8.84	8.60 †	7.07	5.88	9.06	3.43 †	7.37
FFHFA Transverse	-4.78	7.51	-4.78	7.17	-7.00	7.01	-4.01	7.22
FFTBA Sagittal	-3.22	5.31	-3.51	5.01	-2.95	6.46	-1.79	5.25
FFTBA Frontal	7.33	4.86	7.17	3.75	7.32 ‡	3.77	4.71 ‡	4.74
FFTBA Transverse	2.59	4.45	6.16 †	4.64	1.39	4.49	2.75 †	5.17
HFTBA Sagittal	-8.81	6.56	-9.40	6.76	-10.87	6.76	-11.17	5.47
HFTBA Frontal	-0.37	8.26	-3.26	8.81	-0.51	9.43	0.08	6.55
HFTBA Transverse	6.70	6.26	10.26	5.80	7.63	7.83	6.56	7.12
HIP Sagittal	-3.66	8.34	-5.57	5.66	-6.02	5.97	-6.13	5.97
HIP Frontal	3.95	9.59	5.46	8.98	4.80	11.25	5.99	9.35
HIP Transverse	17.61	18.88	13.04	16.01	12.67	17.41	12.50	14.24
KNEE Sagittal	-10.67	2.86	-10.66	2.66	-12.00	4.45	-12.00	4.34
KNEE Frontal	1.89	2.21	1.69	2.03	1.41	3.21	1.36	3.33
KNEE Transverse	-5.61	6.30	-5.55	5.53	-4.72	5.33	-5.25	5.40
TRUNK Sagittal	12.45	5.65	12.91	5.66	10.64	7.55	10.59	7.25
TRUNK Frontal	0.84	1.28	-0.61	1.59	0.59	1.82	0.03	2.03
TRUNK Transverse	-5.39	3.65	-3.73	2.94	-4.93	3.20	-4.23	4.57

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Table F.3 Peak angular displacement from heel strike to 200 ms post initial contact (degrees)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	11.35	8.45	9.63	9.01	12.42	9.97	12.85	7.66
	Plantarflexion	5.78	7.84	5.22	8.98	7.28	10.23	8.23	7.98
FFHFA Frontal	Inversion	9.57	7.89	7.22	8.74	6.56	9.49	4.26	7.52
	Eversion	4.27	7.86	1.71	8.73	2.11	9.23	-0.24	7.54
FFHFA Transverse	Adduction	-4.63	7.10	-4.45	7.57	-6.24	7.13	-3.39	7.41
	Abduction	-9.13	6.82	-8.74	7.41	-10.17	7.19	-7.60	7.72
FFTBA Sagittal	Dorsiflexion	2.95	4.52	2.55	4.50	2.36	5.74	3.44	5.43
	Plantarflexion	-7.14	4.43	-6.83	4.80	-7.03	6.19	-5.87	5.16
FFTBA Frontal	Inversion	7.17	3.74	7.33	4.86	7.33 ‡	3.76	4.88 ‡	4.49
	Eversion	-0.36	2.60	-0.95	4.59	0.04 ‡	3.89	-2.21 ‡	4.18
FFTBA Transverse	Internal Rotation	6.27*	4.74	2.74	4.47	1.72*	4.72	3.25	5.38
	External Rotation	0.03	5.48	-3.23	4.75	-2.77	5.55	-1.99	5.63
HFTBA Sagittal	Dorsiflexion	-5.83	5.87	-5.64	5.94	-7.94	6.42	-7.92	5.35
	Plantarflexion	-12.65	6.37	-12.60	7.35	-15.29	6.34	-14.78	5.35
HFTBA Frontal	Inversion	-3.10	8.75	0.00	7.79	-0.29	9.31	0.39	6.49
	Eversion	-8.05	9.87	-5.21	7.73	-5.29	9.40	-4.45	7.01
HFTBA Transverse	Internal Rotation	11.01	5.88	7.31	6.10	8.64	7.93	7.56	7.10
	External Rotation	6.93	5.85	3.40	6.23	4.58	7.86	3.44	7.47
HIP Sagittal	Extension	32.84	5.15	32.97	5.21	33.63	8.00	33.79	7.17
	Flexion	20.49	5.56	20.87	6.12	21.18	8.05	20.83	7.02
HIP Frontal	Adduction	4.07	2.65	5.72	2.91	4.47	3.65	4.17	3.79
	Abduction	-2.70	2.49	-1.46	2.93	-1.80	2.49	-2.19	4.40
HIP Transverse	Internal Rotation	-1.12	7.24	-1.30	8.18	-1.26	6.21	-3.07	8.52
	External Rotation	-6.98	6.96	-7.89	7.32	-6.99	4.46	-8.07	7.77
KNEE Sagittal	Extension	-10.61	2.63	-10.67	2.86	-12.00	4.45	-11.99	4.36
	Flexion	-22.61	4.04	-22.41	3.93	-23.23	6.31	-23.40	6.59
KNEE Frontal	Adduction	2.46	2.00	2.83	2.12	2.92	3.36	2.39	3.13
	Abduction	1.09	1.85	1.34	2.22	1.21	3.21	0.79	3.21
KNEE Transverse	Internal Rotation	-2.98	5.34	-3.76	5.94	-2.85	5.19	-2.74	4.95
	External Rotation	-6.15	5.28	-6.40	5.96	-5.53	5.25	-6.03	5.25
TRUNK Sagittal	Extension	13.78	5.77	13.59	5.96	11.57	7.87	11.43	7.46
	Flexion	12.36	5.85	12.07	5.85	9.96	7.55	10.01	7.17
TRUNK Frontal	Adduction	4.07	2.08	5.73 †	1.60	5.09	2.70	4.22 †	2.27
	Abduction	-0.61*	1.59	0.84	1.28	0.59*	1.82	-0.23	2.02
TRUNK Transverse	Internal Rotation	-1.72	2.47	-3.06	3.22	-2.47	2.68	-2.94	2.95
	External Rotation	-4.04	2.80	-5.70	3.69	-5.05	3.13	-5.53	2.99

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Angular Velocity

Table F.4 Peak angular velocity from 100 ms pre-initial contact to heel strike(degrees/second)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	38.51	37.96	24.94	31.38	32.39	32.53	34.14	34.26
	Plantarflexion	-65.23	37.26	-55.20	35.97	-57.58	19.44	-60.33	35.97
FFHFA Frontal	Inversion	43.47	19.07	44.21	33.73	43.87	20.73	43.83	25.63
	Eversion	-31.34	24.51	-38.05	36.18	-28.29	18.46	-40.55	23.41
FFHFA Transverse	Adduction	37.33	18.28	27.43	20.70	34.95	21.03	32.80	17.41
	Abduction	-35.38	20.31	-42.45	22.96	-30.42	19.99	-37.63	13.79
FFTBA Sagittal	Dorsiflexion	9.99*	18.60	-1.70	24.03	-2.20*	13.24	0.51	21.40
	Plantarflexion	-87.79	30.49	-83.54	25.17	-84.33	22.53	-79.84	33.96
FFTBA Frontal	Inversion	35.97	27.64	35.12	21.04	42.62	19.74	34.25	23.29
	Eversion	-75.76	28.45	-81.30	27.57	-70.44	35.35	-73.04	29.19
FFTBA Transverse	Internal Rotation	38.24	19.66	33.74	22.01	32.40	15.58	30.32	23.36
	External Rotation	-32.21	21.11	-43.63 †	18.97	-26.44	21.35	-29.90 †	20.93
HFTBA Sagittal	Dorsiflexion	26.36	41.14	10.26	15.93	22.29	20.52	20.06	26.49
	Plantarflexion	-76.91	28.09	-67.40	38.35	-82.08	37.56	-73.69	28.31
HFTBA Frontal	Inversion	36.61	24.79	38.76	29.22	35.97	26.08	43.46	25.12
	Eversion	-76.60	23.15	-83.30	33.42	-71.19	30.43	-78.80	33.35
HFTBA Transverse	Internal Rotation	36.38	22.93	35.98	31.07	39.41	20.38	39.32	22.65
	External Rotation	-30.86	21.73	-28.71	23.84	-37.47	19.82	-32.33	21.44
HIP Sagittal	Extension	11.35	12.93	14.03	16.94	9.18	26.63	10.67	25.52
	Flexion	-29.83	17.88	-31.99	21.48	-29.42	18.71	-28.25	21.14
HIP Frontal	Adduction	36.99	14.88	33.02	13.44	34.07	17.94	30.05	12.04
	Abduction	-7.36	16.88	-7.03	14.52	-14.33	12.12	-12.07	13.15
HIP Transverse	Internal Rotation	56.91	25.44	60.87	27.77	47.32	34.10	44.26	30.09
	External Rotation	-62.05	32.59	-60.30	32.27	-44.34	21.66	-43.58	26.13
KNEE Sagittal	Extension	267.85	51.91	277.17	53.81	266.02	49.59	264.31	54.17
	Flexion	-137.50	27.10	-131.01	35.99	-120.73	48.80	-128.94	52.17
KNEE Frontal	Adduction	19.38	14.54	20.32	14.94	19.96	19.30	17.17	10.91
	Abduction	-17.39	11.32	-18.57	16.93	-22.14	23.05	-18.14	18.42
KNEE Transverse	Internal Rotation	68.03	36.83	65.38	36.86	89.25	29.60	84.81	31.39
	External Rotation	-36.81	21.61	-39.90	22.77	-34.71	20.54	-39.58	23.86
TRUNK Sagittal	Extension	14.61	7.03	13.39	8.10	12.59	11.19	12.51	9.05
	Flexion	-4.66	9.51	-3.16	6.61	-5.68	7.73	-4.89	6.37
TRUNK Frontal	Adduction	29.54	12.69	29.64	9.77	24.85	8.79	25.77	11.80
	Abduction	-1.70	5.41	-2.65	5.54	-2.98	7.65	-2.52	6.52
TRUNK Transverse	Internal Rotation	5.86	8.99	6.44	8.69	5.80	7.44	4.30	6.06
	External Rotation	-18.80	9.83	-20.55	9.63	-22.14	7.67	-22.32	7.32

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Table F.5 Angular velocity at heel strike(degrees/second)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	-13.57	23.54	-20.48	58.05	-15.29	35.49	-18.98	50.73
FFHFA Frontal	-4.08	36.74	13.96	26.37	0.33	28.27	4.80	34.46
FFHFA Transverse	-27.62*	19.21	-20.84	26.72	-6.39*	28.21	-7.67	23.34
FFTBA Sagittal	-70.18	32.73	-62.39	41.87	-75.09	30.45	-73.70	37.14
FFTBA Frontal	-81.13	27.65	-72.23	32.78	-69.82	35.66	-69.01	34.67
FFTBA Transverse	-35.61*	25.53	-25.87	26.30	-9.71*	28.78	-14.73	32.58
HFTBA Sagittal	-56.21	37.35	-41.08	50.98	-60.99	47.06	-54.04	35.76
HFTBA Frontal	-71.90	28.66	-74.04	25.75	-64.84	28.97	-67.63	43.79
HFTBA Transverse	-12.19	19.83	1.00	29.79	-1.84	27.08	-5.91	29.11
HIP Sagittal	5.62	18.69	4.62	13.51	-1.50	27.57	-1.27	30.22
HIP Frontal	8.94	18.30	11.72	22.99	3.26	21.20	5.45	18.94
HIP Transverse	58.01	28.44	52.23	27.10	41.43	34.67	40.41	32.84
KNEE Sagittal	-130.72	35.93	-136.53	27.53	-120.29	48.71	-128.60	52.50
KNEE Frontal	-4.25*	13.76	-4.73 †	11.40	6.95*	12.09	3.73 †	11.94
KNEE Transverse	-6.92	30.97	-8.73	24.11	-13.33	25.65	-9.92	25.10
TRUNK Sagittal	9.64	10.28	10.44	8.79	7.35	14.42	6.36	11.69
TRUNK Frontal	-3.31	31.58	2.17	32.76	8.52	25.43	-6.41	28.20
TRUNK Transverse	-1.72	10.93	0.42	11.03	2.03	9.48	-2.37	7.25

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Table F.6 Peak angular velocity from heel strike to 200 ms post-initial contact (degrees/second)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	67.91	22.59	59.60	26.47	59.13	20.12	59.78	22.60
	Plantarflexion	-57.48	46.85	-41.46	28.11	-46.67	25.18	-57.83	40.63
FFHFA Frontal	Inversion	37.36	28.53	34.74	25.32	29.74	17.75	35.66	20.70
	Eversion	-60.05	17.66	-73.28 †	30.76	-51.04	21.12	-53.52 †	21.72
FFHFA Transverse	Adduction	23.05	14.22	21.19	11.64	28.96	19.79	27.71	14.63
	Abduction	-62.55	15.20	-62.89	21.69	-57.41	15.64	-59.88	19.46
FFTBA Sagittal	Dorsiflexion	98.08	19.22	90.20	18.94	90.57	18.29	91.68	23.00
	Plantarflexion	-93.03	45.36	-95.37	35.20	-102.48	28.01	-102.46	30.37
FFTBA Frontal	Inversion	8.91	11.40	10.35	8.74	4.00 ‡	6.02	10.57 ‡	11.48
	Eversion	-96.72	36.24	-109.81	30.39	-89.34	36.13	-92.56	40.27
FFTBA Transverse	Internal Rotation	34.53	23.00	31.63	19.32	37.59	15.65	38.34	22.09
	External Rotation	-80.86*	21.96	-79.18	28.38	-60.41*	20.60	-69.41	25.36
HFTBA Sagittal	Dorsiflexion	79.67	22.20	75.14	25.50	84.81	16.13	83.10	26.14
	Plantarflexion	-75.67	36.25	-88.99	40.67	-98.00	39.41	-86.20	40.99
HFTBA Frontal	Inversion	24.78	14.71	25.37	20.03	20.12	8.82	24.00	13.68
	Eversion	-89.63	33.39	-89.68	31.60	-83.35	27.22	-81.66	38.03
HFTBA Transverse	Internal Rotation	46.50	23.37	43.36	19.58	49.85	15.54	49.30	21.85
	External Rotation	-57.16	19.72	-54.44	22.21	-53.63	21.88	-52.08	22.49
HIP Sagittal	Extension	5.42	13.17	9.95	17.74	0.71	25.59	0.84	27.10
	Flexion	-109.08	18.54	-108.55	17.84	-111.02	9.41	-111.08	9.52
HIP Frontal	Adduction	43.33	19.49	46.24	20.34	39.36	14.62	41.93	18.45
	Abduction	-23.48	12.34	-25.72	14.03	-22.00	13.15	-21.51	9.76
HIP Transverse	Internal Rotation	100.40	33.14	103.37	37.51	93.51	31.34	87.02	27.97
	External Rotation	-29.58	23.06	-23.81	21.49	-35.80	24.03	-29.61	23.38
KNEE Sagittal	Extension	58.80	26.31	54.96	21.31	53.81	13.98	58.59	18.87
	Flexion	-144.39	29.01	-142.53	33.49	-131.35	40.93	-138.93	47.88
KNEE Frontal	Adduction	16.98	7.27	16.97	7.69	22.41	14.16	18.47	12.38
	Abduction	-11.81	6.77	-12.09	7.64	-10.08	9.72	-10.73	7.94
KNEE Transverse	Internal Rotation	53.11	27.78	46.98	21.83	44.26	18.91	50.68	28.54
	External Rotation	-35.27	13.11	-38.84	18.07	-39.22	18.75	-38.60	15.08
TRUNK Sagittal	Extension	15.91	7.55	17.38	8.10	15.50	10.67	15.47	8.89
	Flexion	-13.18	6.57	-10.87	7.23	-14.56	6.55	-14.12	6.69
TRUNK Frontal	Adduction	45.81	15.49	48.89	15.59	42.64	13.47	41.75	13.85
	Abduction	-32.10	12.57	-30.33	11.47	-30.43	10.16	-29.18	10.18
TRUNK Transverse	Internal Rotation	27.22	8.16	29.77	10.07	30.46	7.50	28.02	5.21
	External Rotation	-1.67	6.82	-1.36	8.04	-2.68 ‡	5.04	0.35 ‡	5.23

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Angular Acceleration

Table F.7 Peak angular acceleration from 100 ms pre-initial contact to heel strike(degrees/second^2)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	+ VE	1987.60	1167.02	1821.22	1139.98	2011.69	1237.57	1774.85	936.40
	- VE	-2025.12	1791.10	-1625.72	1197.75	-1810.90	692.85	-2031.94	1187.50
FFHFA Frontal	+ VE	1498.06	1017.79	1459.55	1171.95	1450.59	1024.81	1670.51	610.70
	- VE	-1559.17	940.11	-2073.86	1834.45	-1600.34	752.33	-1859.96	917.56
FFHFA Transverse	+ VE	1089.64	676.49	1188.52	1114.29	1213.80	732.31	1399.70	928.45
	- VE	-1633.17	746.68	-1621.92	605.04	-1489.32	557.09	-1410.85	517.95
FFTBA Sagittal	+ VE	1509.21	812.14	1380.50	893.95	1347.21	418.81	1214.65	836.86
	- VE	-2371.81	1128.62	-2289.70	1054.85	-2520.74	644.72	-2525.85	762.56
FFTBA Frontal	+ VE	446.88	600.31	353.50	615.48	237.84	610.49	589.42	677.49
	- VE	-2325.13	920.27	-2447.17	792.87	-2147.05	852.76	-2215.29	1065.25
FFTBA Transverse	+ VE	600.92*	443.11	807.51	429.67	906.03*	400.22	916.17	570.22
	- VE	-1552.77	410.54	-1838.24	706.27	-1343.76	699.39	-1394.00	630.33
HFTBA Sagittal	+ VE	1757.29	1655.48	1157.48	752.14	1516.88	902.88	1362.47	1060.02
	- VE	-2470.20	1123.28	-2069.32	1013.04	-2734.46	1354.45	-2220.84	912.15
HFTBA Frontal	+ VE	730.40	902.06	1166.39	1947.66	724.68	577.89	947.34	926.03
	- VE	-2363.80	803.92	-2533.87	1089.24	-2459.15	1008.29	-2312.51	718.82
HFTBA Transverse	+ VE	1222.99	500.50	1078.68	730.22	1311.41	613.41	1280.29	522.31
	- VE	-1523.20	755.17	-1749.01	1263.23	-1703.93	840.85	-1608.79	788.69
HIP Sagittal	+ VE	785.31	316.24	834.34	446.41	725.39	465.12	673.56	473.94
	- VE	-440.14	221.29	-452.22	301.97	-527.43	253.48	-498.86	295.52
HIP Frontal	+ VE	562.47	432.03	552.06	414.49	611.32	527.50	618.85	334.74
	- VE	-937.86	421.12	-836.12	392.37	-959.79	495.84	-923.30	433.56
HIP Transverse	+ VE	2249.84	671.98	2106.90	725.14	1806.01	704.66	1775.11	436.07
	- VE	12.29	472.56	-54.33	513.10	-315.60	590.55	-142.07	461.24
KNEE Sagittal	+ VE	-610.15	794.46	-773.79	729.16	-573.30	606.36	-646.05	591.05
	- VE	-5879.80	836.50	-5947.33	1078.66	-5713.26	1230.20	-5767.31	1414.65
KNEE Frontal	+ VE	395.23	394.32	437.55	414.09	562.17	556.52	524.54	517.36
	- VE	-583.56*	381.81	-592.92	532.40	-326.44*	339.92	-361.43	320.38
KNEE Transverse	+ VE	1324.99	801.10	1567.37	909.15	1107.34	749.41	1382.83	1078.71
	- VE	-2407.78	1111.72	-2420.18	1078.65	-2738.85	959.92	-2831.50	1084.70
TRUNK Sagittal	+ VE	338.89	160.28	321.52	122.99	294.55	146.74	282.51	191.97
	- VE	-182.44	140.06	-135.81	141.57	-153.67	160.21	-178.44	125.98
TRUNK Frontal	+ VE	525.29	156.55	526.02	184.42	480.60	166.40	475.91	181.70
	- VE	25.04	158.58	32.89	85.76	37.73	108.57	37.10	118.18
TRUNK Transverse	+ VE	446.12	229.57	500.37	226.07	478.98	148.07	460.61	143.35
	- VE	-20.65	122.77	-18.95	110.50	29.58	121.17	25.97	96.07

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Table F.8 Angular acceleration at heel strike (degrees/second^2)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	-817.83	792.86	-688.23	1300.48	-521.65	864.99	-709.06	1103.74
FFHFA Frontal	-1014.27	1506.55	-525.16	1221.11	-378.25	939.94	-419.90	1134.83
FFHFA Transverse	-528.30	817.44	-310.00	543.42	-237.40	835.50	-256.74	772.50
FFTBA Sagittal	-1709.21	999.04	-1934.35	1192.63	-1905.02	778.99	-1942.29	676.24
FFTBA Frontal	-1698.97 *	908.53	-1517.03	1087.94	-883.42 *	770.10	-1074.14	1195.05
FFTBA Transverse	-506.46	1315.12	-463.70	540.84	-157.65	790.86	-310.93	902.65
HFTBA Sagittal	-1055.07	1070.54	-1315.82	863.76	-1529.43	808.43	-1292.79	1214.17
HFTBA Frontal	-513.47	1517.48	-756.07	927.74	-378.51	743.04	-491.04	789.24
HFTBA Transverse	-190.93	1090.59	-152.07	600.00	-14.56	831.38	-192.93	979.82
HIP Sagittal	-209.53	477.96	-319.29	324.31	-344.86	342.02	-351.46	342.06
HIP Frontal	226.13	549.29	313.11	514.29	275.20	644.36	342.92	535.58
HIP Transverse	1008.79	1081.54	747.39	917.16	725.85	997.46	716.12	816.10
KNEE Sagittal	-784.63	754.85	-610.15	794.46	-606.73	653.85	-680.48	642.24
KNEE Frontal	-16.84	150.55	-34.31	176.39	-67.60	213.28	-56.31	178.38
KNEE Transverse	1006.02	1006.29	1056.74	818.72	790.28	856.30	996.88	1231.05
TRUNK Sagittal	114.27	191.28	102.61	221.81	67.75	174.70	44.99	222.92
TRUNK Frontal	30.55	436.43	-4.56	406.67	165.18	379.90	-137.50	358.12
TRUNK Transverse	-95.17	460.52	-7.55	377.20	30.25	438.23	-58.57	402.34

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Table F.9 Peak angular acceleration from heel strike to 200 ms post-initial contact (degrees/second^2)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	+ VE	2444.21	777.33	2058.24	992.14	2078.06	574.46	2645.16	1318.05
	- VE	-1871.56	630.44	-1764.27	999.89	-1879.01	745.98	-1867.87	740.85
FFHFA Frontal	+ VE	1540.56	358.34	1734.25	725.31	1491.80	494.96	1548.12	601.22
	- VE	-2162.25	626.18	-2524.09	1291.96	-1886.72	837.19	-2036.99	774.35
FFHFA Transverse	+ VE	1478.80	355.69	1714.32	552.15	1577.91	607.48	1599.63	551.38
	- VE	-1476.75	633.52	-1447.21	455.23	-1747.05	766.75	-1669.65	763.91
FFTBA Sagittal	+ VE	3759.51	1166.36	3677.13	944.39	3662.75	738.90	3809.95	1095.61
	- VE	-2248.01	831.17	-1924.79	693.89	-2043.63	636.38	-2081.60	645.94
FFTBA Frontal	+ VE	1579.76	589.32	1817.96	541.38	1457.84	657.53	1669.93	663.30
	- VE	-1704.37	882.56	-1828.85	739.45	-1225.43	549.69	-1528.99	829.67
FFTBA Transverse	+ VE	2215.39	766.46	2195.87	838.85	1904.41 ‡	697.36	2207.67 ‡	848.05
	- VE	-1642.67	783.91	-1764.58	693.38	-1722.26	685.18	-1703.49	842.51
HFTBA Sagittal	+ VE	2976.12	807.35	3269.80	1290.37	3480.85	1059.49	3343.48	1138.79
	- VE	-2199.09	687.58	-1973.57	647.49	-2388.20	561.33	-2412.08	1264.18
HFTBA Frontal	+ VE	2114.16	632.67	2441.09	1106.13	2014.45	777.79	2046.18	731.90
	- VE	-1418.53	576.61	-1474.65	624.29	-1467.50	544.46	-1388.59	572.95
HFTBA Transverse	+ VE	2058.08	753.47	2021.60	836.23	2146.13	844.23	2173.92	676.26
	- VE	-1865.91	731.71	-1752.70	591.21	-1806.40	533.17	-1876.90	720.48
HIP Sagittal	+ VE	133.48	215.38	305.45	340.41	300.42	297.54	321.71	303.38
	- VE	-1197.04	295.55	-1388.41	718.26	-1272.70	348.35	-1261.10	362.72
HIP Frontal	+ VE	863.40	262.76	1051.38	886.41	963.43	436.51	926.53	399.77
	- VE	-979.64	351.52	-1052.33	490.31	-934.04	250.64	-980.27	278.62
HIP Transverse	+ VE	1566.27	760.78	1535.94	683.41	1575.74	584.11	1457.86	553.60
	- VE	-2316.29	1144.58	-2286.64	1038.06	-2059.47	834.22	-1952.50	713.66
KNEE Sagittal	+ VE	2135.07	458.12	2100.92	378.98	2058.42	483.36	2124.03	534.73
	- VE	-845.00	572.98	-904.44	709.24	-813.69	533.77	-844.67	536.85
KNEE Frontal	+ VE	419.61	184.38	403.78	175.47	465.84	292.04	440.67	268.59
	- VE	-498.68	253.86	-518.78	287.08	-644.95	481.41	-536.17	397.64
KNEE Transverse	+ VE	1944.98	768.97	2013.96	980.45	1830.83	637.09	2024.82	929.69
	- VE	-1892.80	876.70	-2028.09	1028.87	-1728.70	833.54	-1854.65	1011.02
TRUNK Sagittal	+ VE	350.02	149.35	366.40	155.85	374.61	123.75	421.29	157.57
	- VE	-452.73	168.91	-498.21	293.40	-475.74 ‡	194.09	-425.44 ‡	192.41
TRUNK Frontal	+ VE	445.79	181.51	560.88	458.31	458.84	184.58	424.96	147.52
	- VE	-886.06 *	386.39	-961.25	394.73	-554.51 ‡*	224.99	-813.17 ‡	242.27
TRUNK Transverse	+ VE	487.69	141.82	535.00	179.51	539.41	145.32	496.52	150.16
	- VE	-336.59 *	157.01	-393.07	322.26	-479.51 ‡*	149.42	-360.76 ‡	171.36

Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.

Appendix G Study 1 Part 3 – Walk Regressions

SPSS Correlation Outputs

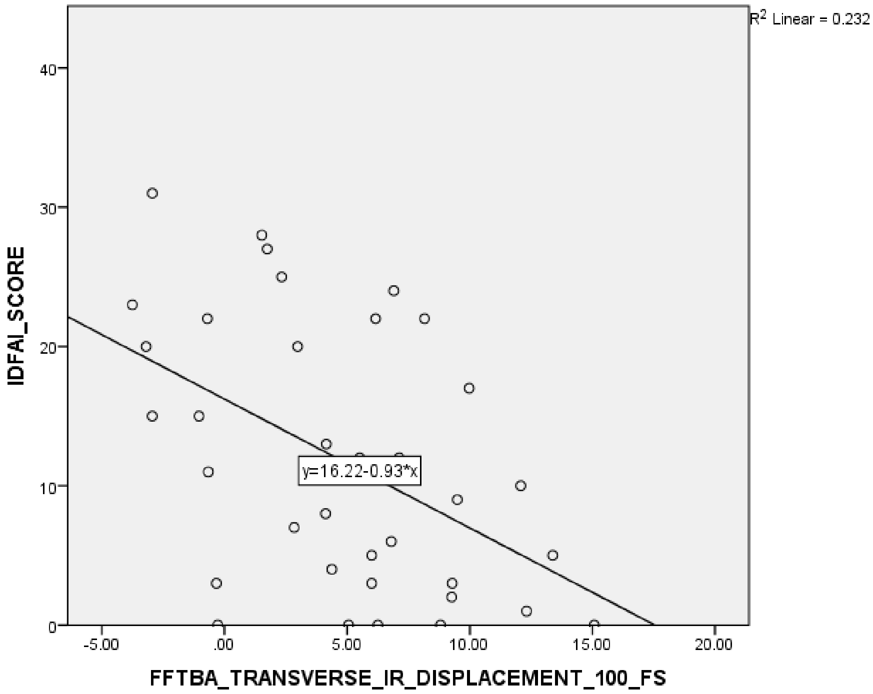


Figure G.1 Pearson's correlation FFTBA transverse internal rotation displacement 100 ms pre HS to HS and IdFAI score

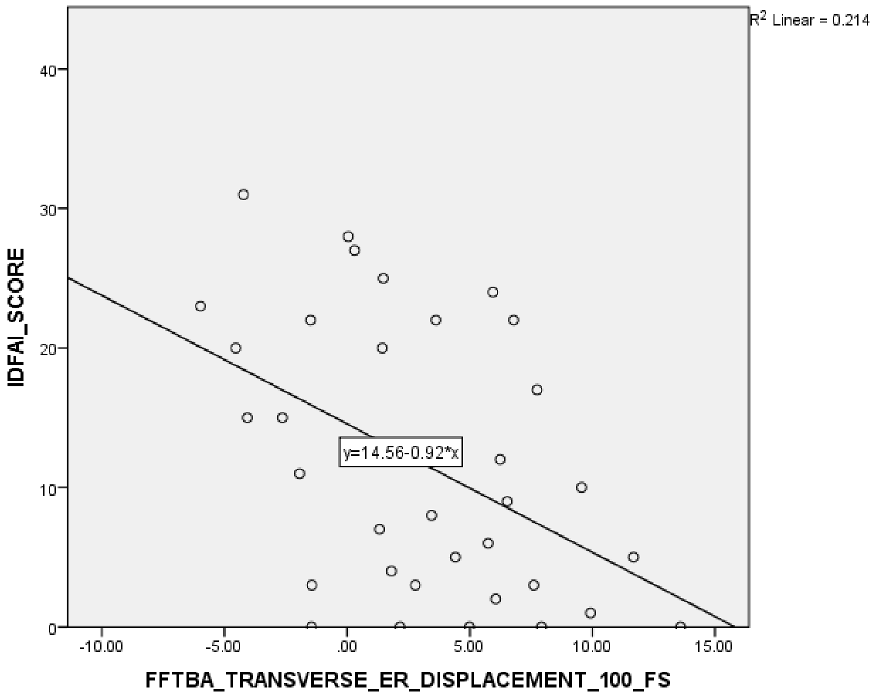


Figure G.2 Pearson's correlation FFTBA transverse external rotation displacement 100 ms pre HS to HS and IdFAI score

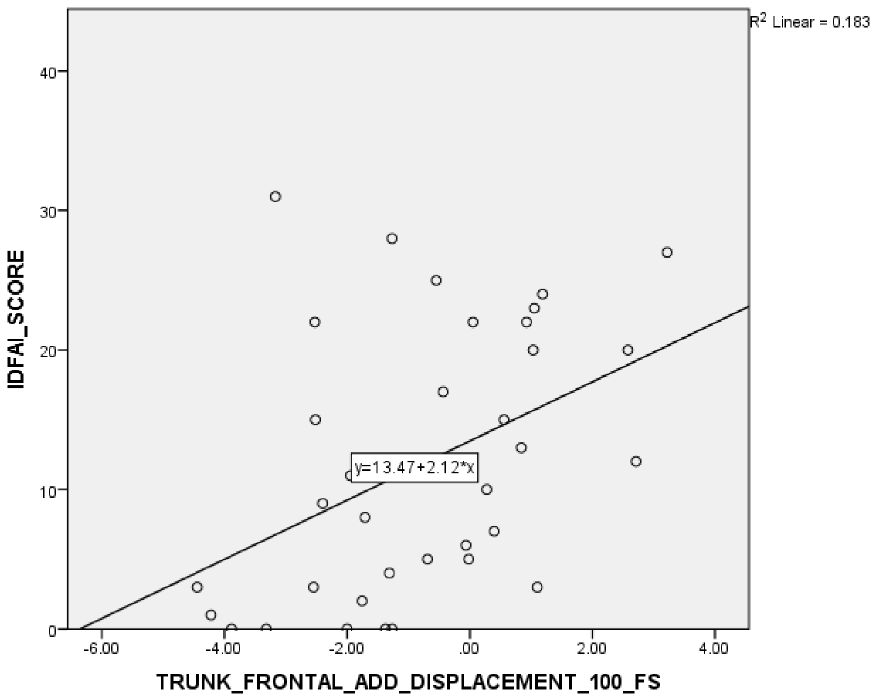


Figure G.3 Pearson's correlation Trunk frontal adduction displacement 100 ms pre HS to HS and IdFAI score

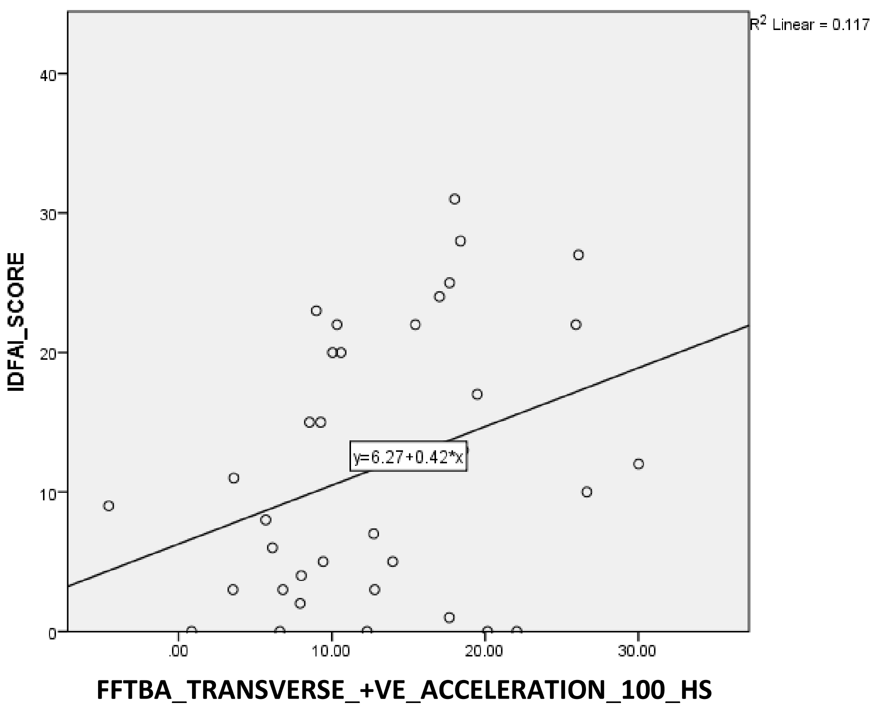


Figure G.4 Pearson's correlation FFTBA transverse internal rotation acceleration 100 ms pre HS to HS and IdFAI score

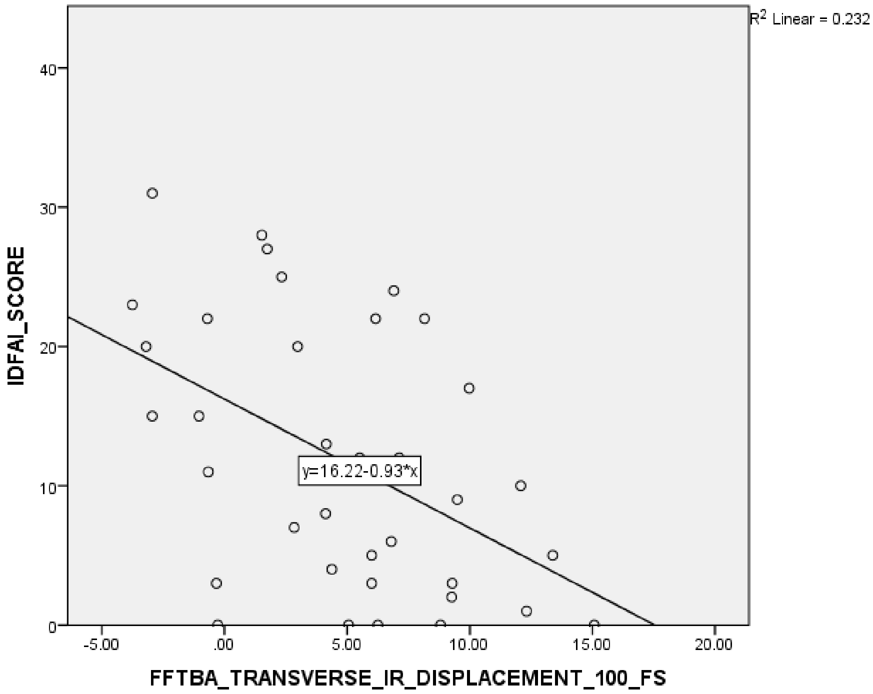


Figure G.5 Pearson's correlation FFTBA transverse internal rotation displacement 100 ms pre HS to HS and IdFAI score

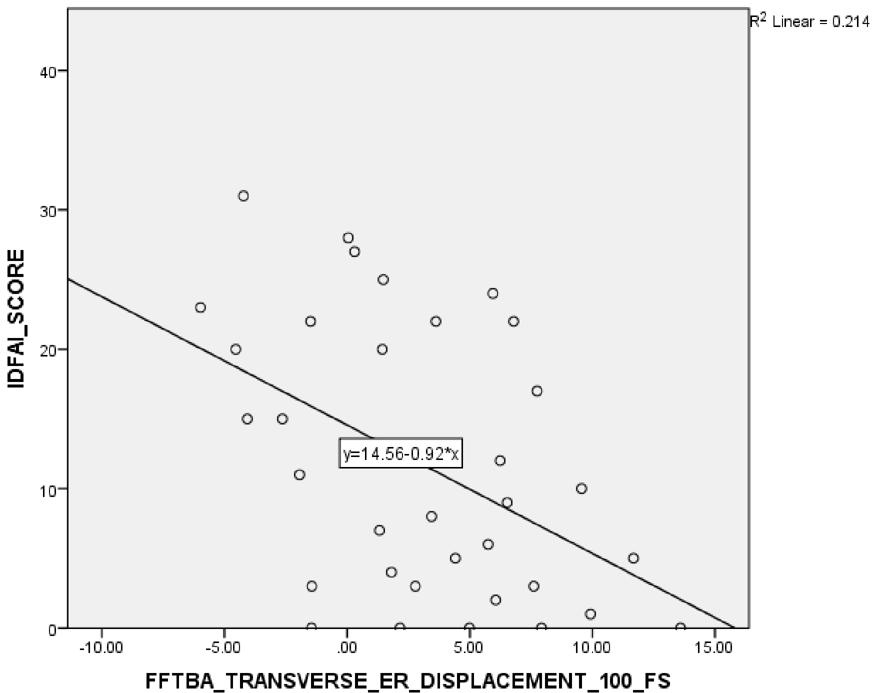


Figure G.6 Pearson's correlation FFTBA transverse external rotation displacement 100 ms pre HS to HS and IdFAI score

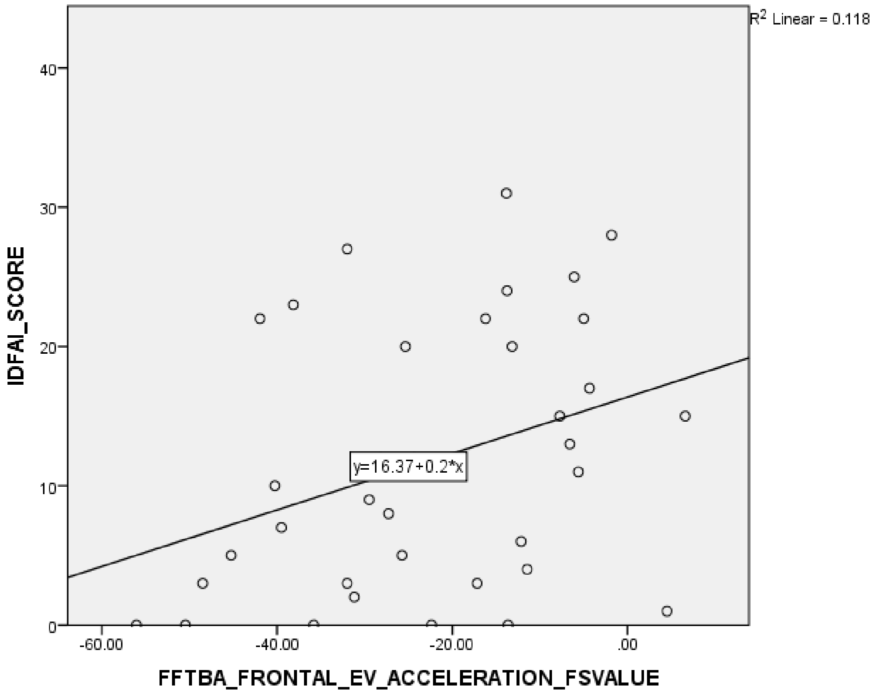


Figure G.7 Pearson's correlation FFTBA frontal eversion acceleration at HS and IdFAI score

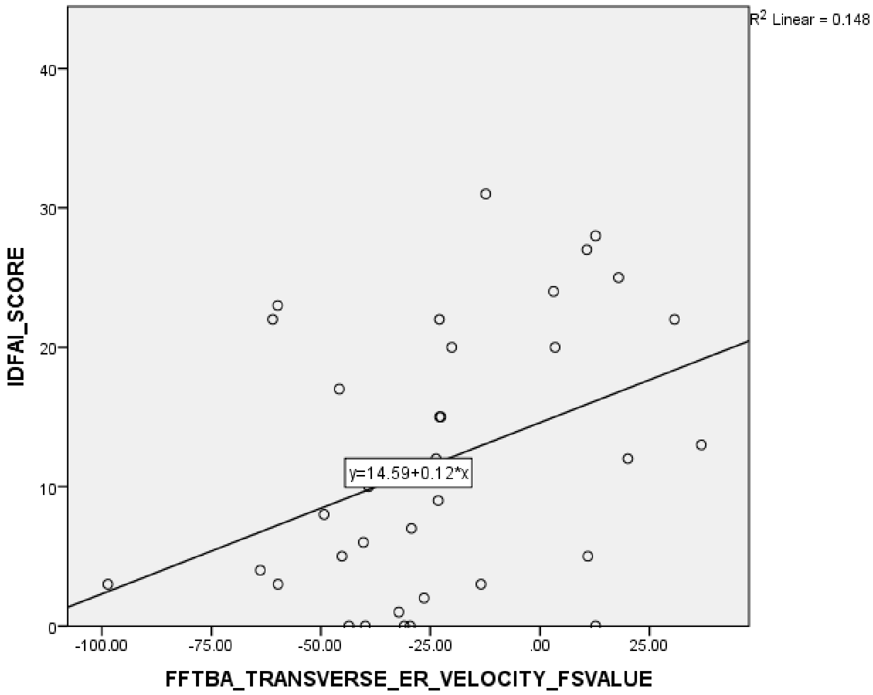


Figure G.8 Pearson's correlation FFTBA transverse external rotation velocity at HS and IdFAI score

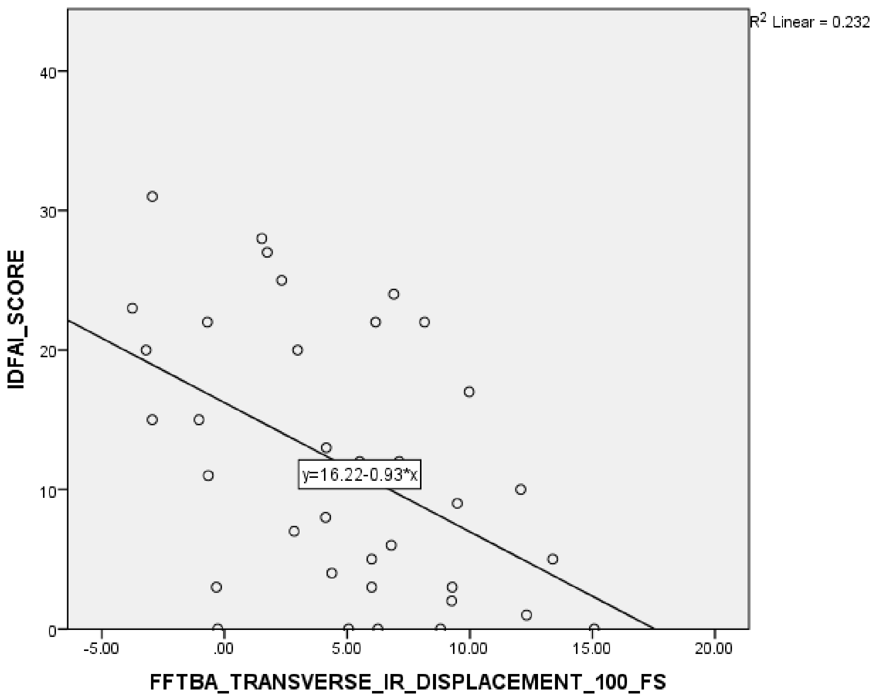


Figure G.9 Pearson's correlation FFTBA transverse internal rotation displacement 100 ms pre HS to HS and IdFAI score

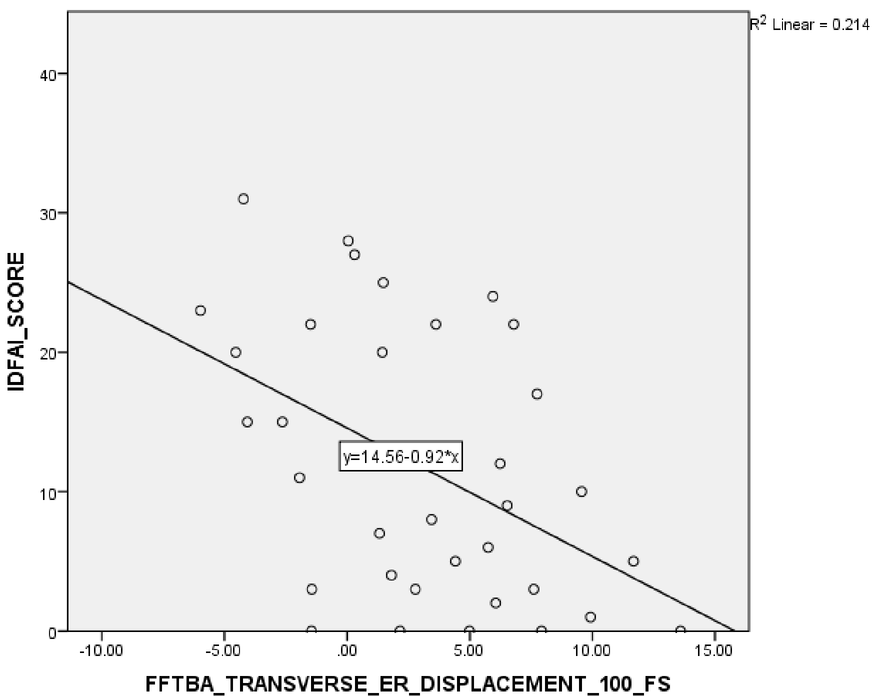


Figure G.10 Pearson's correlation FFTBA transverse external rotation displacement 100 ms pre HS to HS and IdFAI score

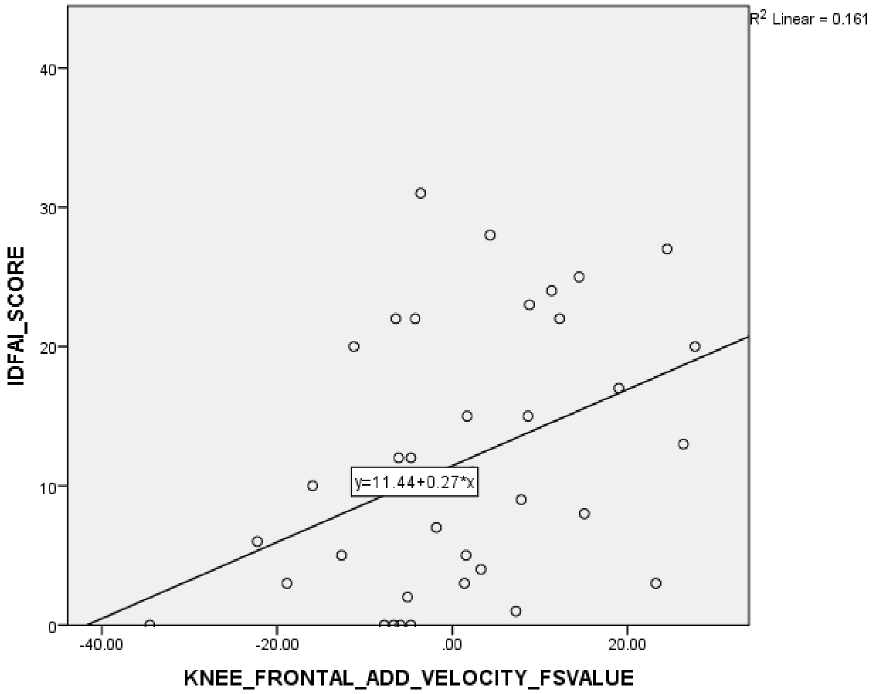


Figure G.11 Pearson's correlation knee frontal adduction velocity at HS and IdFAI score

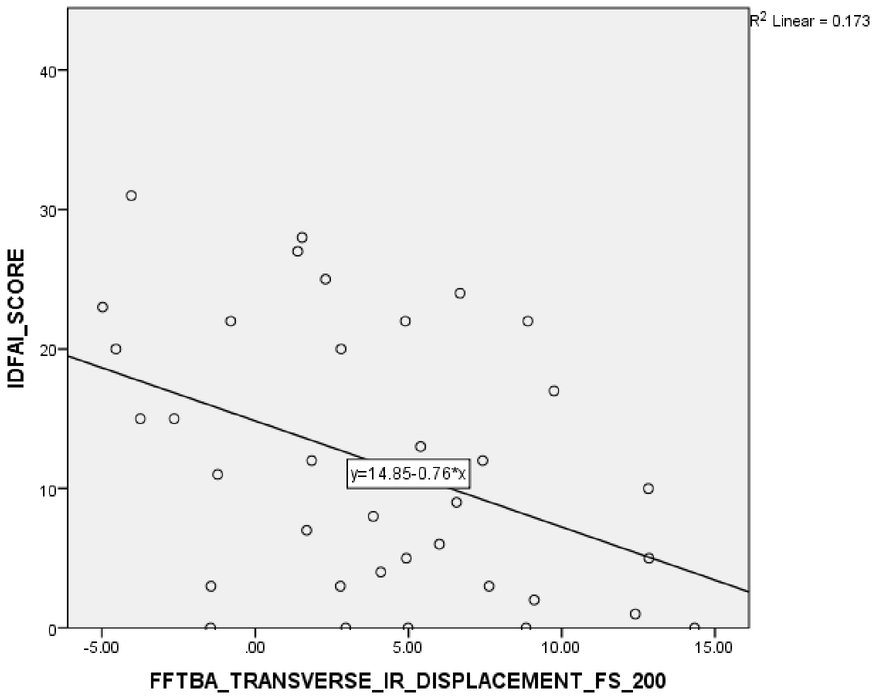


Figure G.12 Pearson's correlation FFTBA transverse internal rotation displacement at HS and IdFAI score

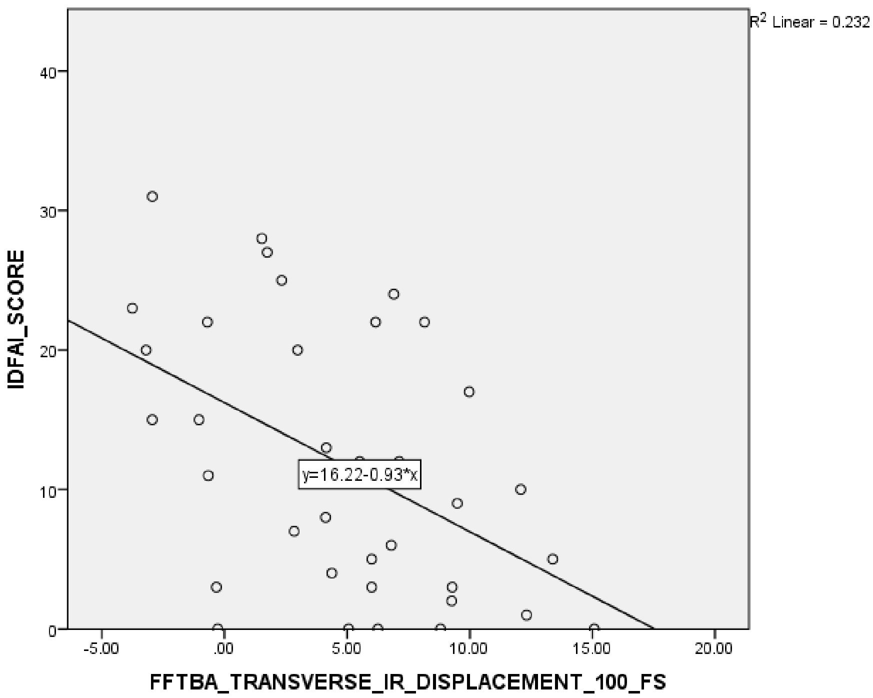


Figure G.13 Pearson's correlation FFTBA transverse internal rotation displacement 100 ms pre HS to HS and IdFAI score

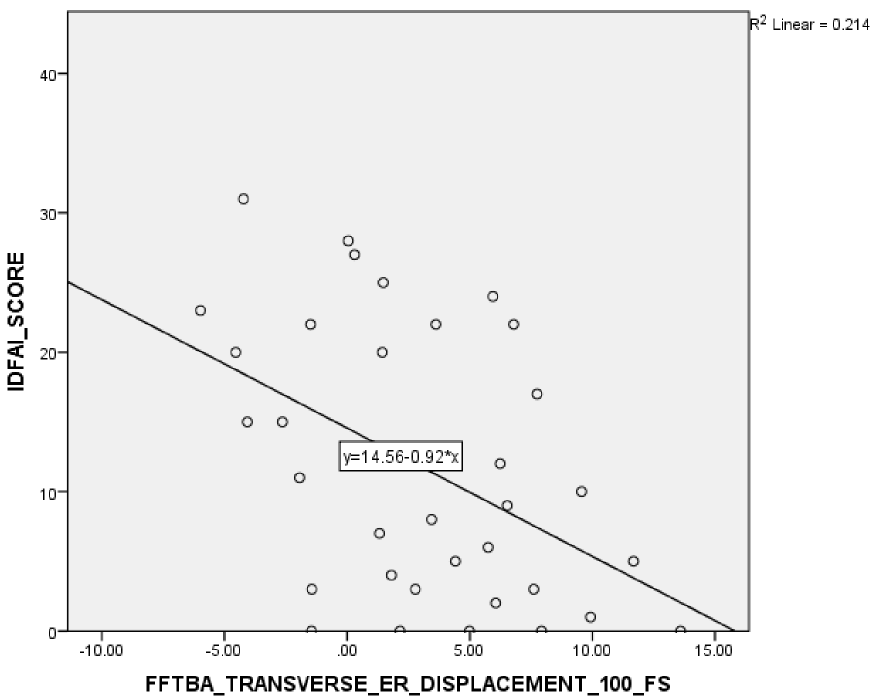


Figure G.14 Pearson's correlation FFTBA transverse external rotation displacement 100 ms pre HS to HS and IdFAI score

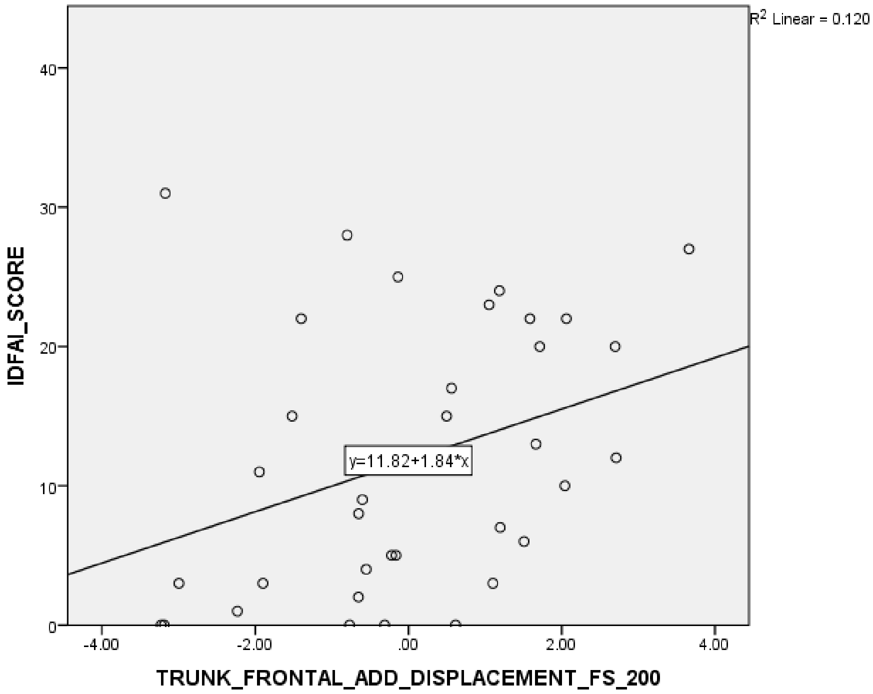


Figure G.15 Pearson's correlation trunk frontal adduction displacement at HS to 200 ms post HS and IdFAI score

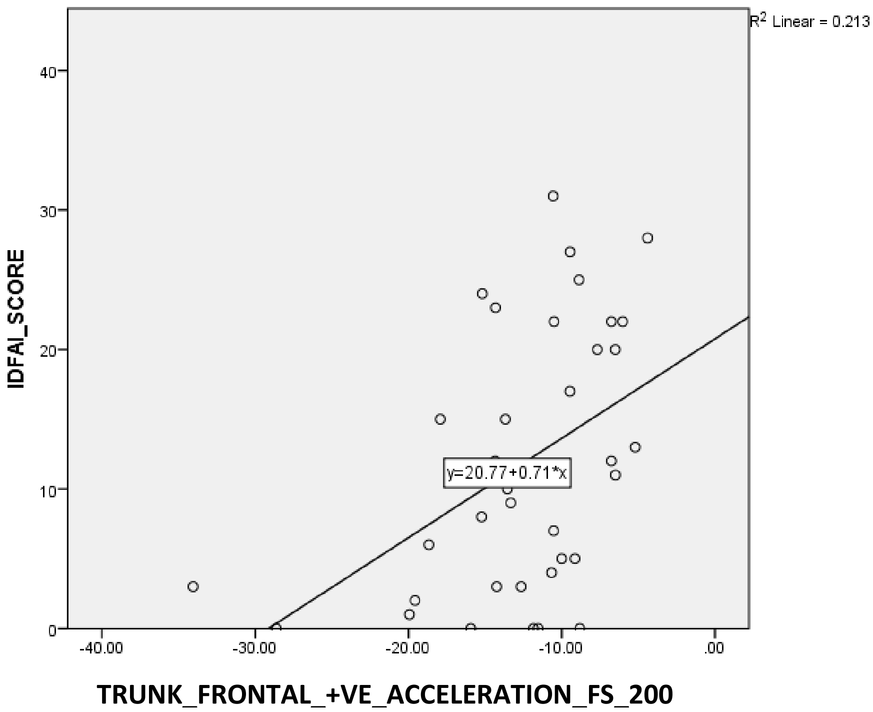


Figure G.16 Pearson's correlation trunk frontal positive acceleration at HS to 200 ms post HS and IdFAI score

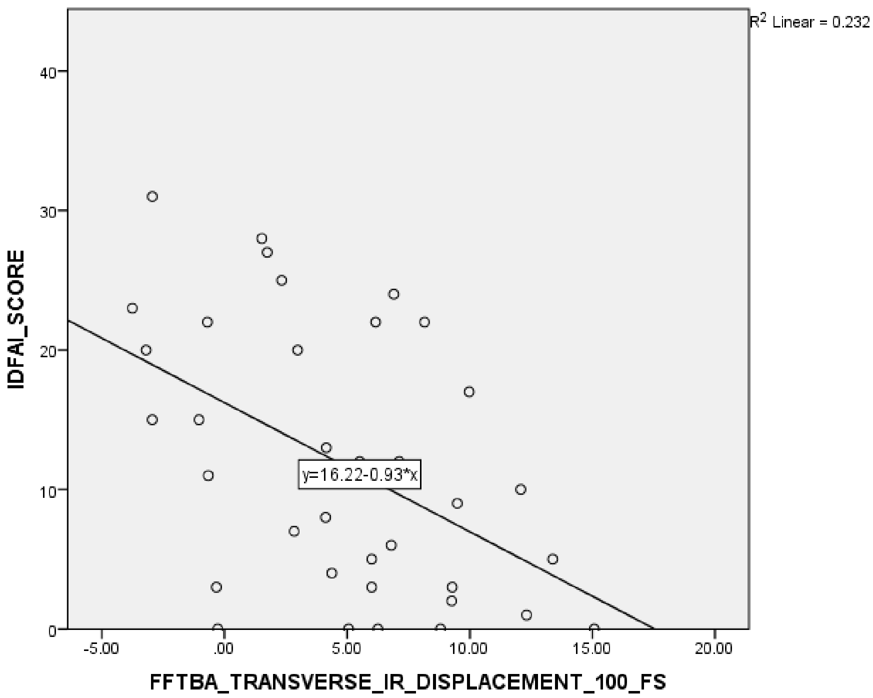


Figure G.17 Pearson's correlation FFTBA transverse internal rotation displacement 100 ms pre HS to HS and IdFAI score

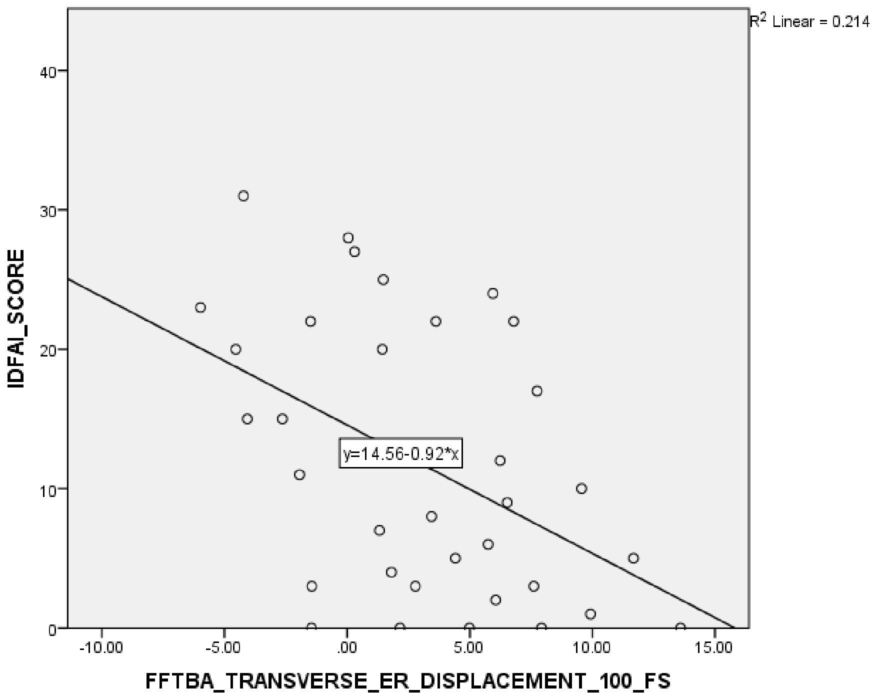


Figure G.18 Pearson's correlation FFTBA transverse external rotation displacement 100 ms pre HS to HS and IdFAI score

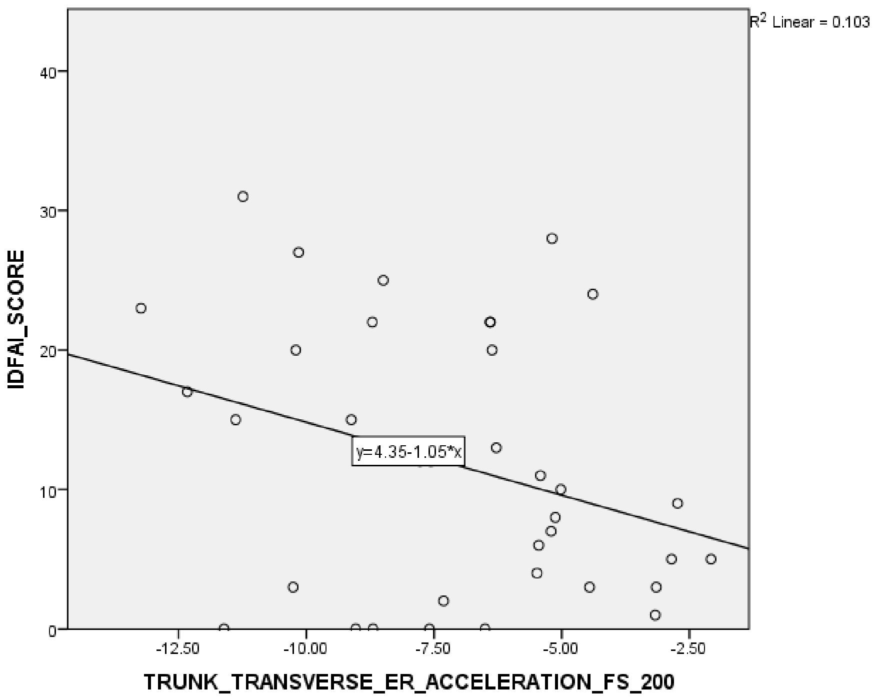


Figure G.19 Pearson's correlation trunk transverse negative acceleration at HS to 200 ms post HS and IdFAI score

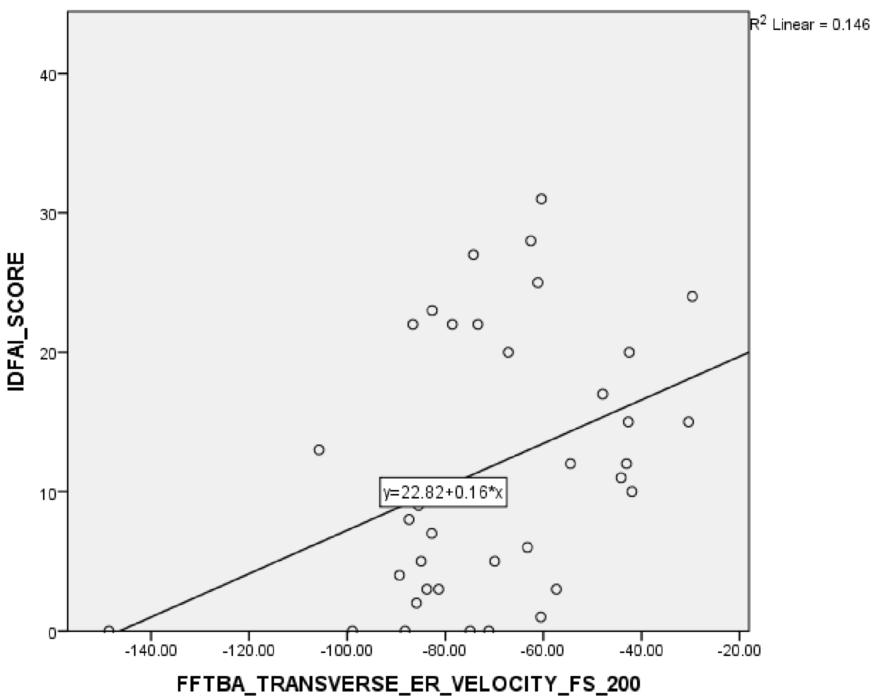


Figure G.20 Pearson's correlation FFTBA transverse external rotation velocity at HS to 200 ms post HS and IdFAI score

Table G.1 R values for regression analysis model with IdFAI as dependent variable

	R	R Square	Adj R Square	F	Sig
FFTBA transverse internal rotation displacement 100 HS	0.482	0.232	0.210	10.285	0.003
FFTBA transverse internal rotation displacement 100 HS, FFTBA transverse +ve acceleration 100 HS	0.648	0.420	0.385	10.683	0.003
FFTBA transverse internal rotation displacement 100 HS, FFTBA transverse +ve acceleration 100 HS, Trunk frontal +ve displacement HS 200	0.705	0.496	0.449	4.858	0.035

Table G.2 Unstandardized and standardized Beta values for each of the 9 regression models

Dependent variable	Variable	B	SE	ß	st error of estimate
IdFAI Score (model 1)	Constant	16.221	1.973		
	FFTBA transverse internal rotation displacement 100 HS	-0.925	0.288	-0.482	8.479
IdFAI Score (model 2)	Constant	9.805	2.624		
	FFTBA transverse internal rotation displacement 100 HS	-1.074	0.259	-0.559	
	FFTBA transverse +ve acceleration 100 HS	0.542	0.166	0.440	7.480
IdFAI Score (model 3)	Constant	12.318	2.732		
	FFTBA transverse internal rotation displacement 100 HS	-1.015	0.246	-0.528	
	FFTBA transverse +ve acceleration 100 HS	0.416	0.167	0.338	
	Trunk frontal +ve displacement HS 200	1.457	0.661	0.294	7.078

Appendix H Study 2 Part 1 – Landing SPM

Matched Control vs Affected limb of CAI group

- **200 ms pre initial contact to initial contact**

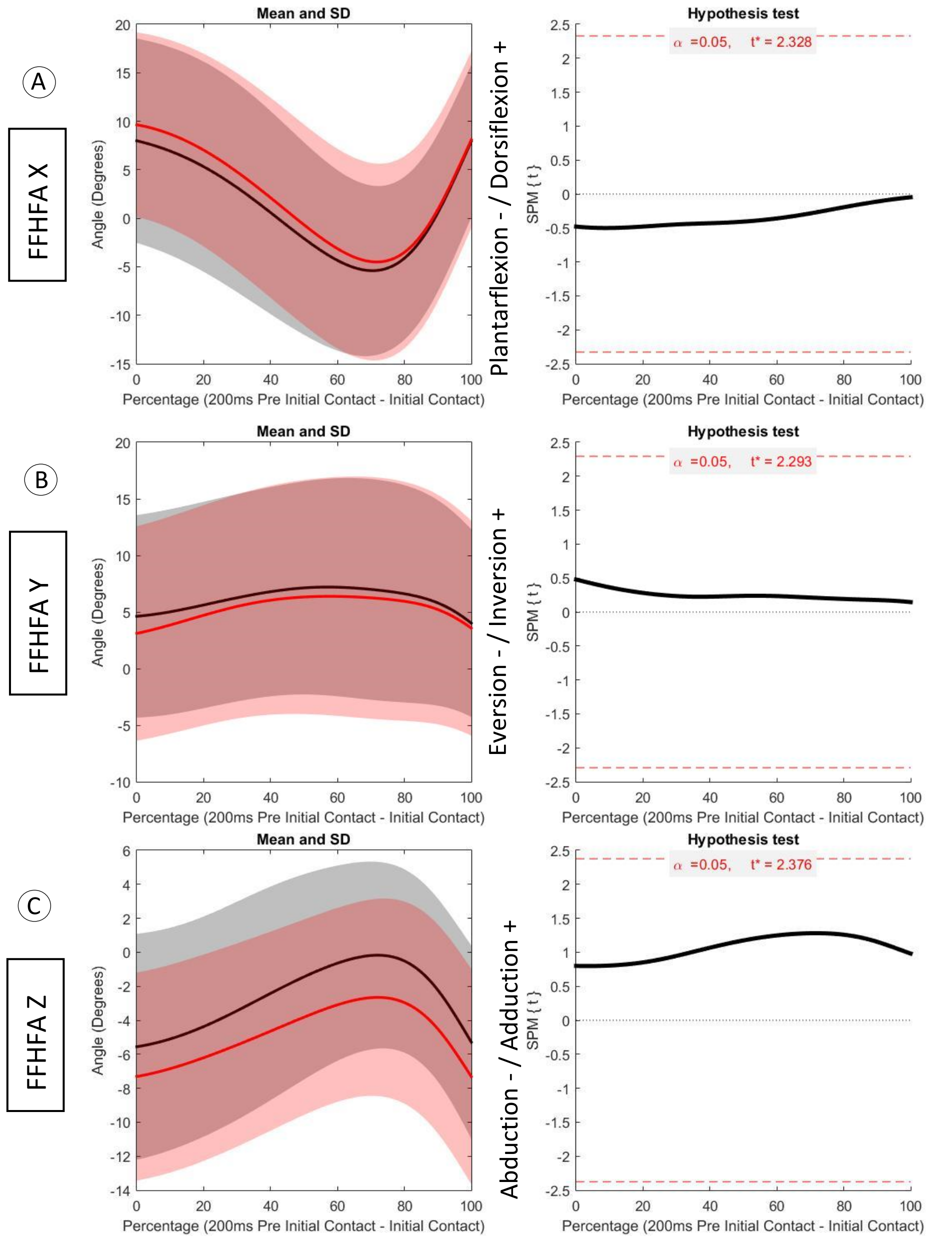


Figure H.1 Single leg landing - Forefoot – Hindfoot angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

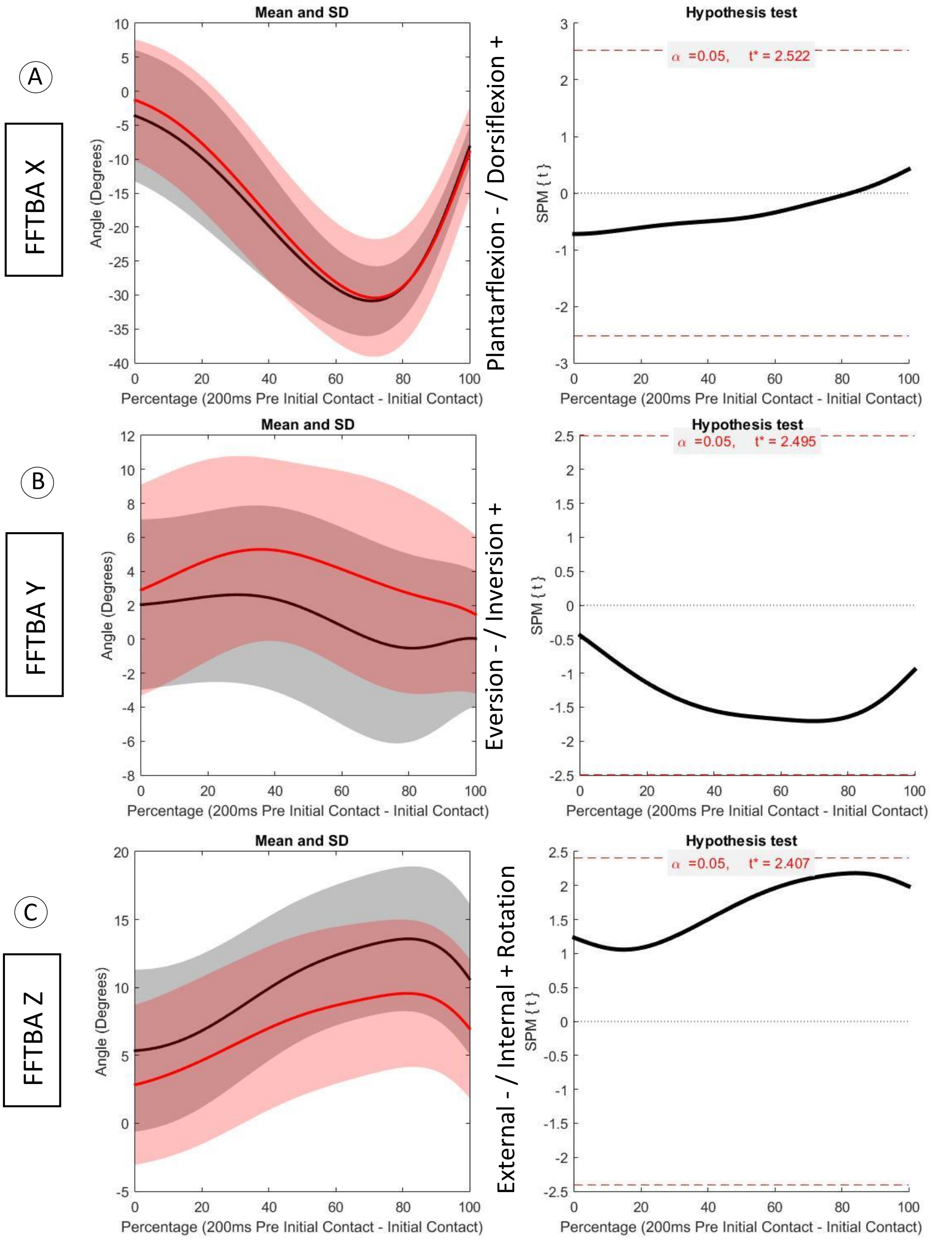


Figure H.2 Single leg landing - Forefoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

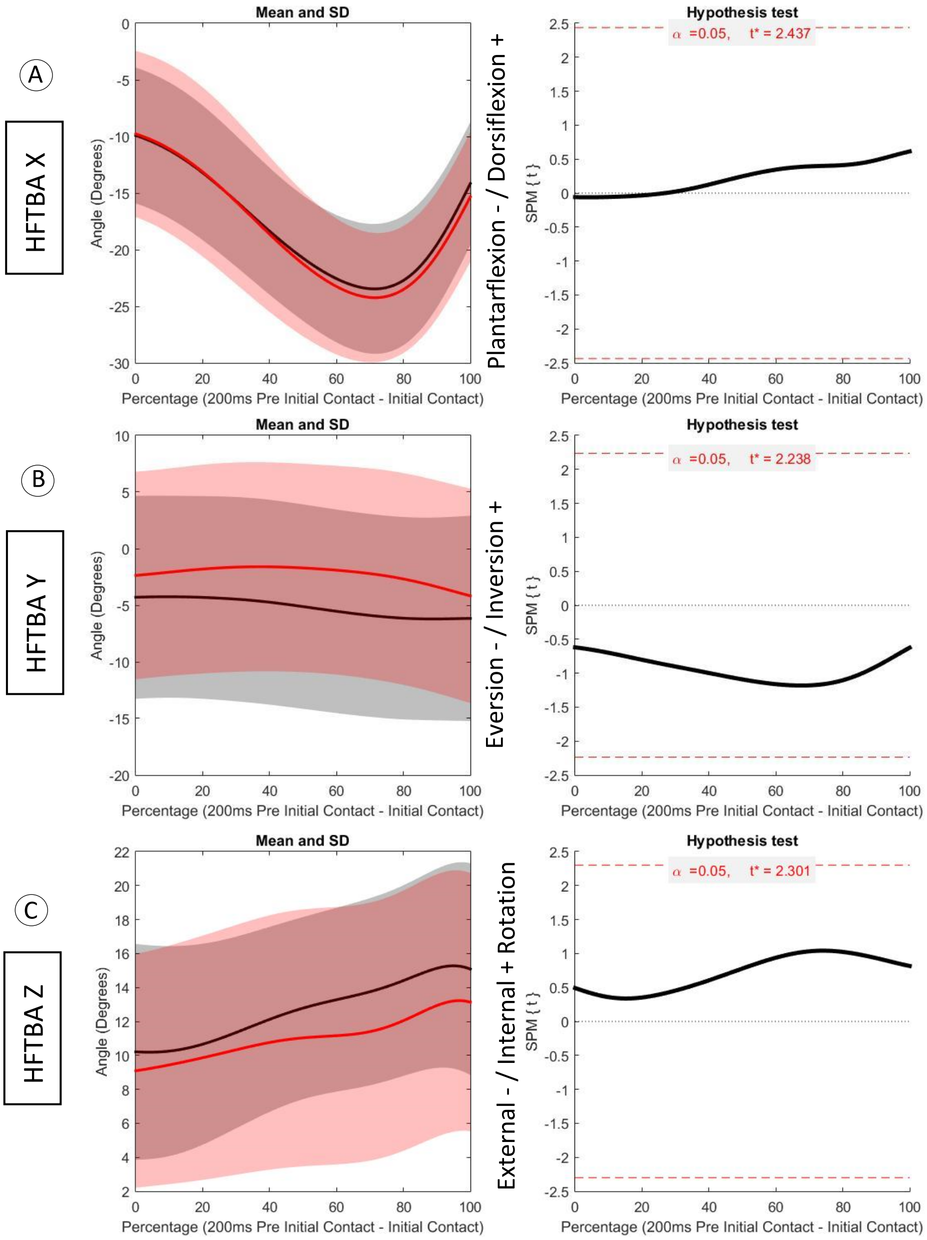


Figure H.3 Single leg landing - Hindfoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

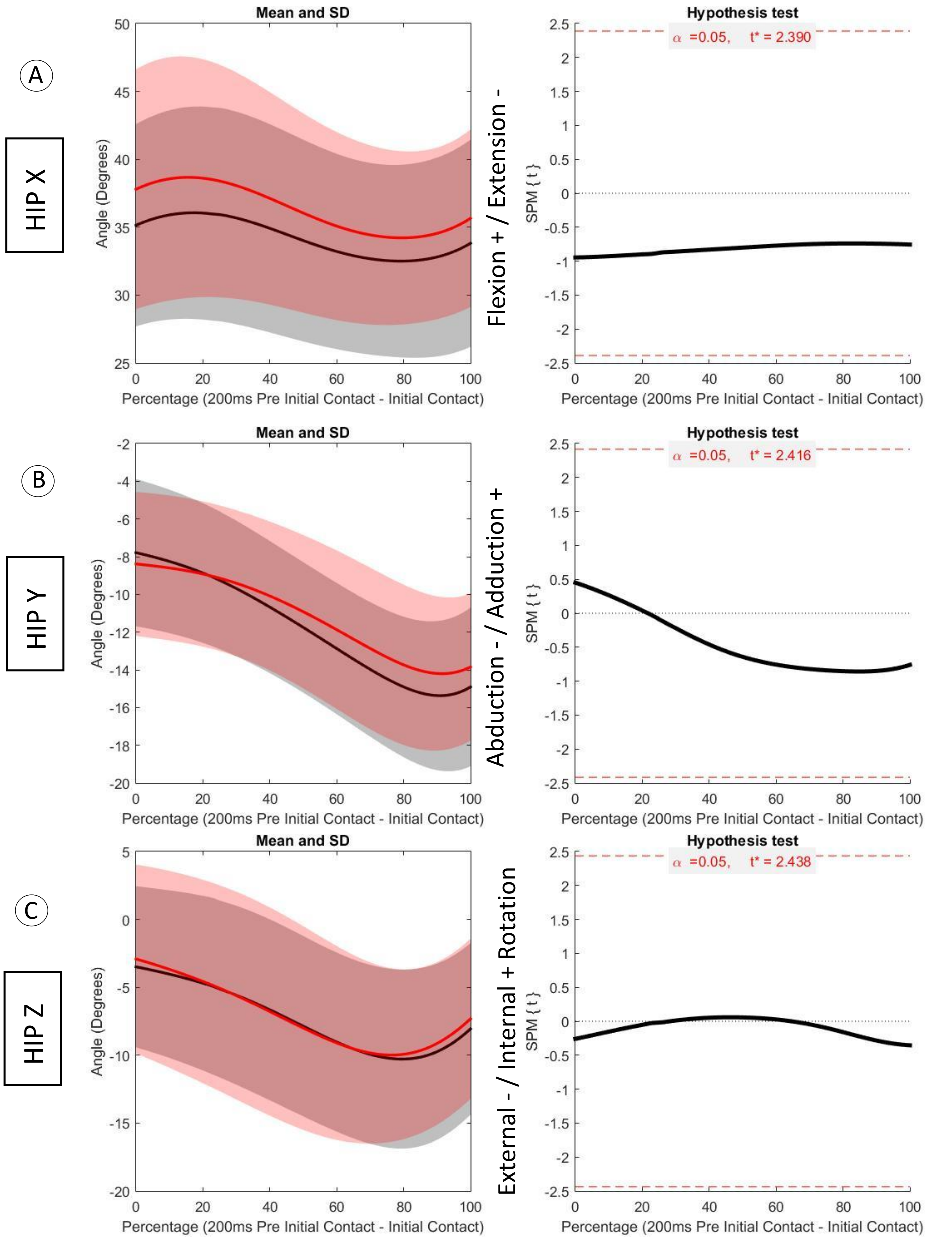


Figure H.4 Single leg landing - Hip angles (x, y, z) 200 ms pre IC to IC - means and standard deviations(Matched Control - — Affected - —) and *t*-test output.

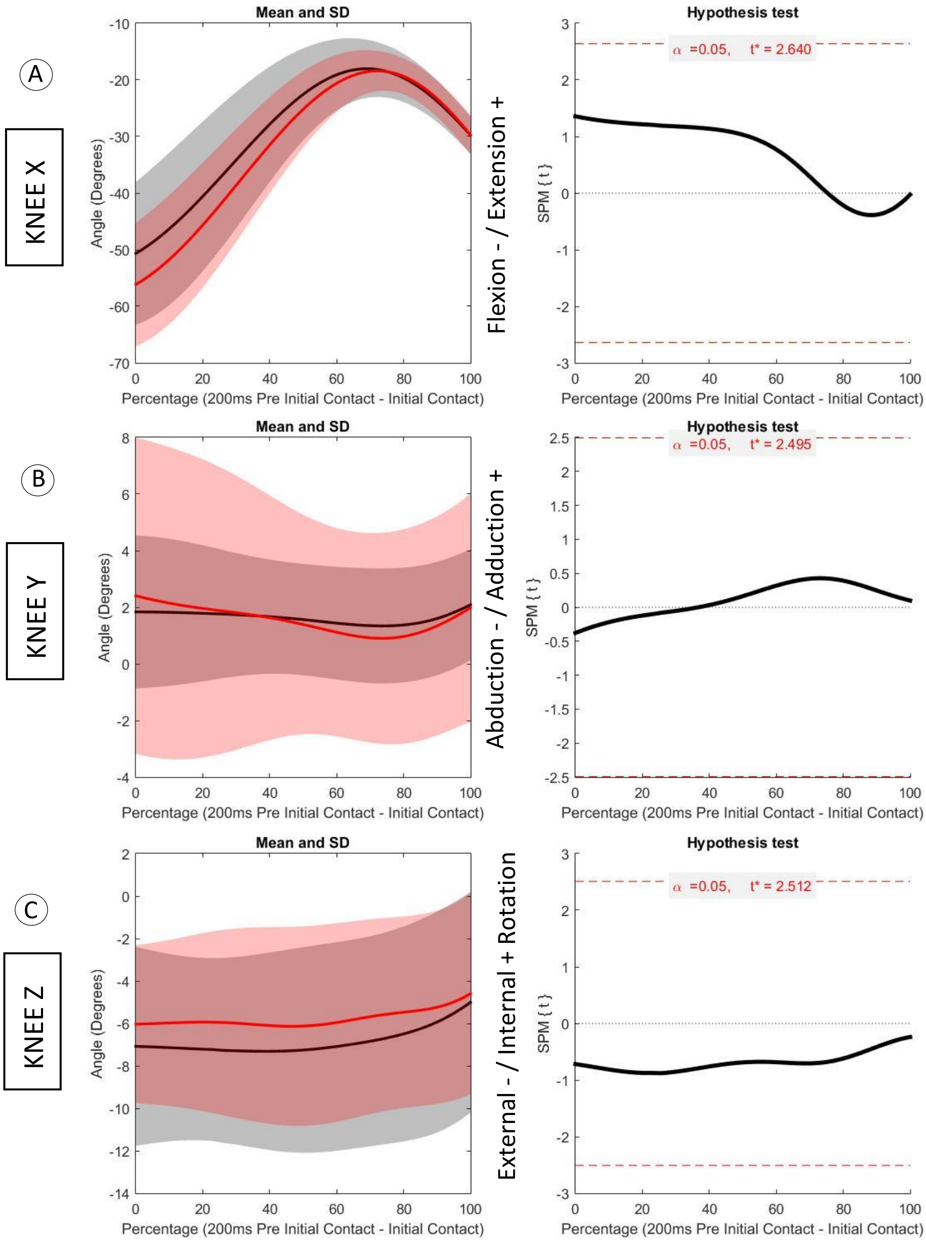


Figure H.5 Single leg landing - Knee angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

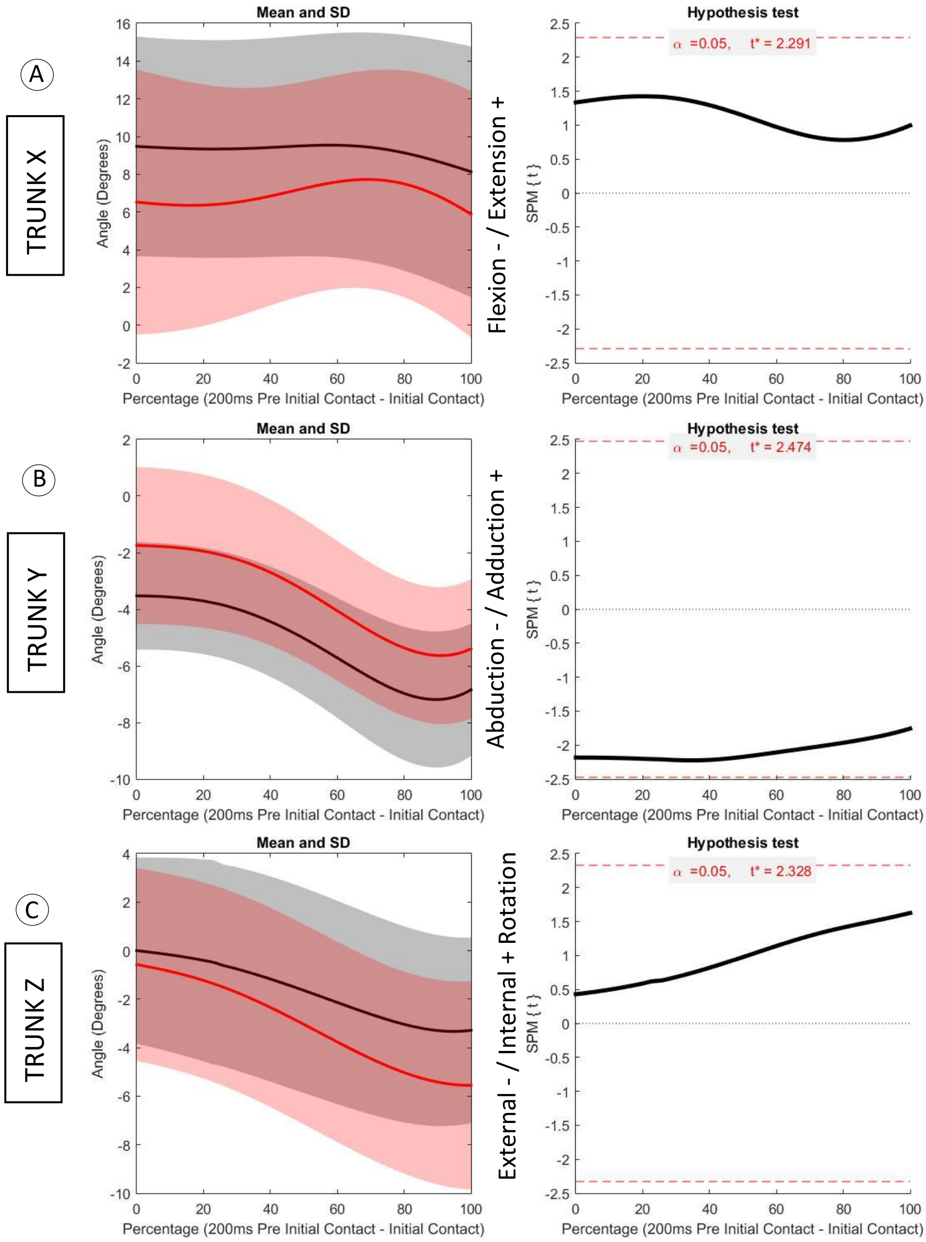


Figure H.6 Single leg landing - Trunk angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

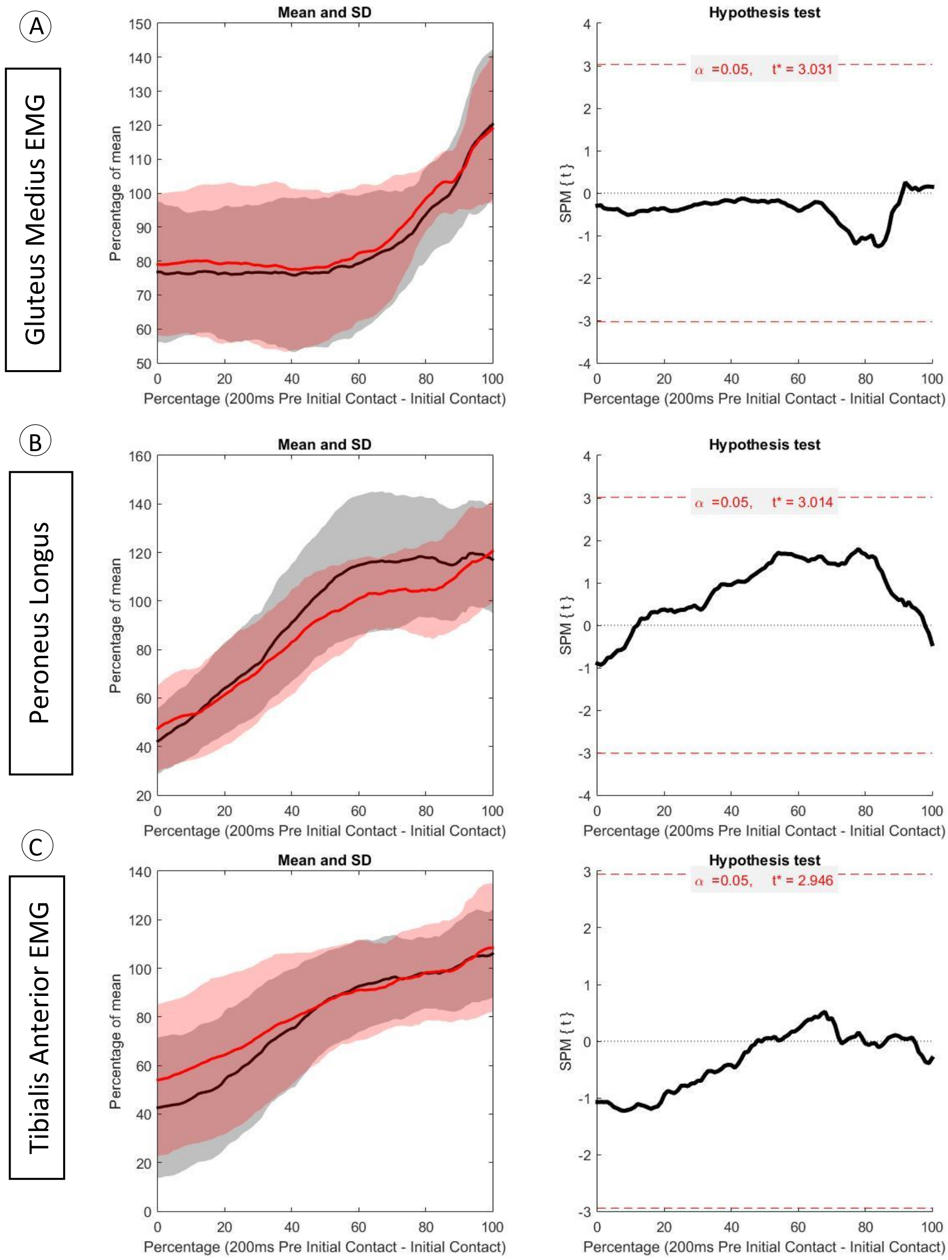


Figure H.7 Single leg landing - Gluteus Medius, Peroneus Longus and Tibialis Anterior muscle activation - 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

- **Initial contact to 200 ms post initial contact**

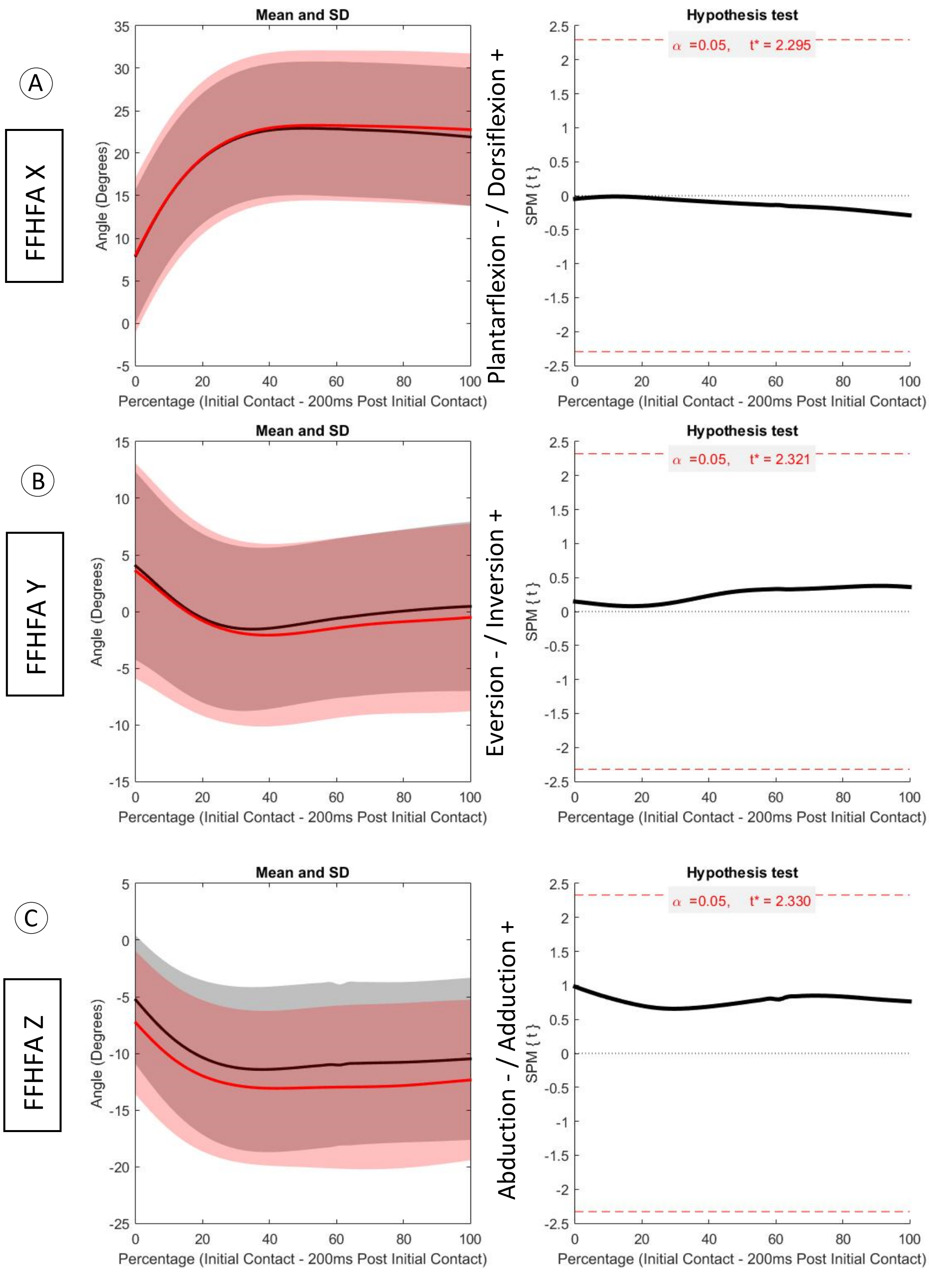


Figure H.8 Single leg landing - Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

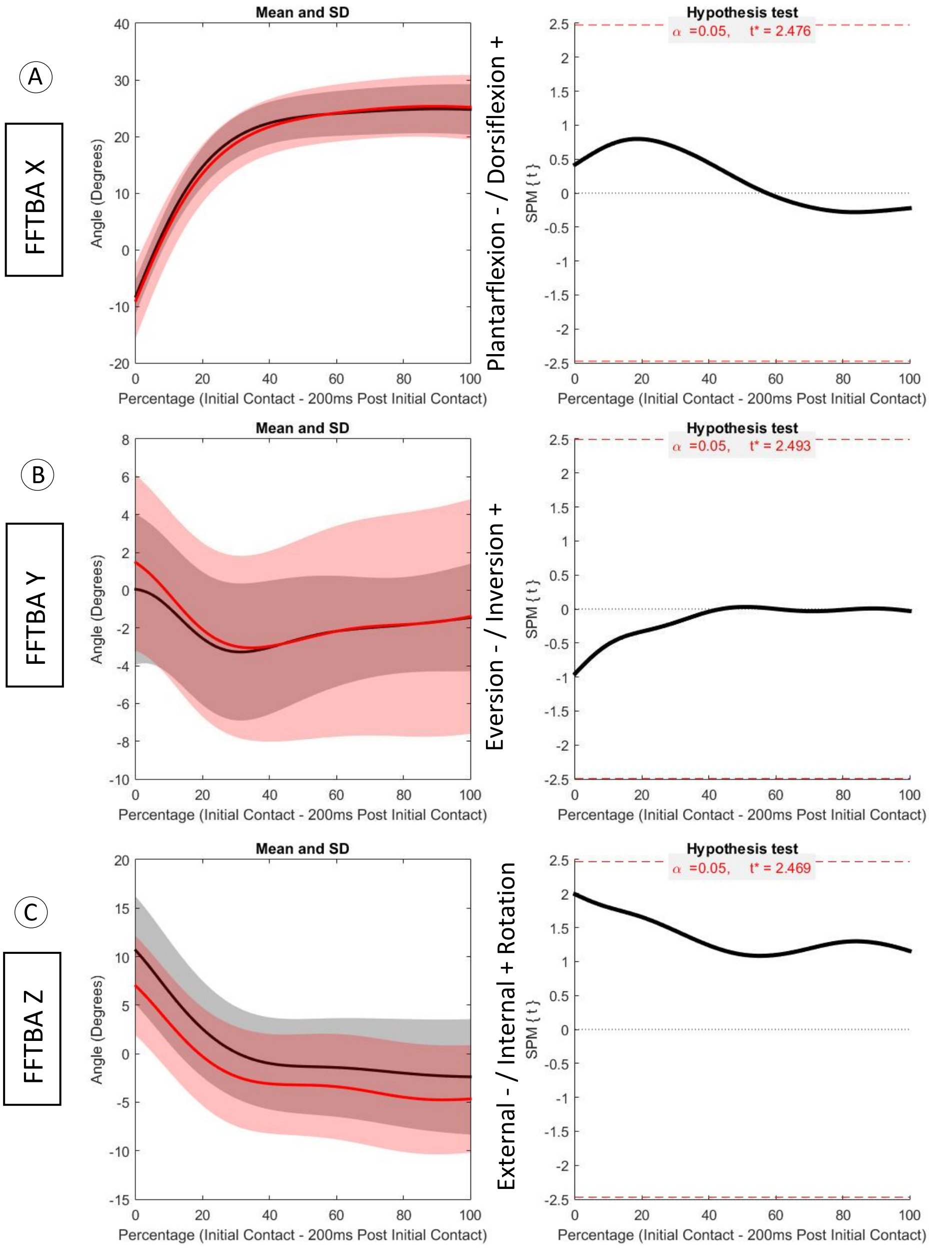


Figure H.9 Single leg landing - Forefoot –Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

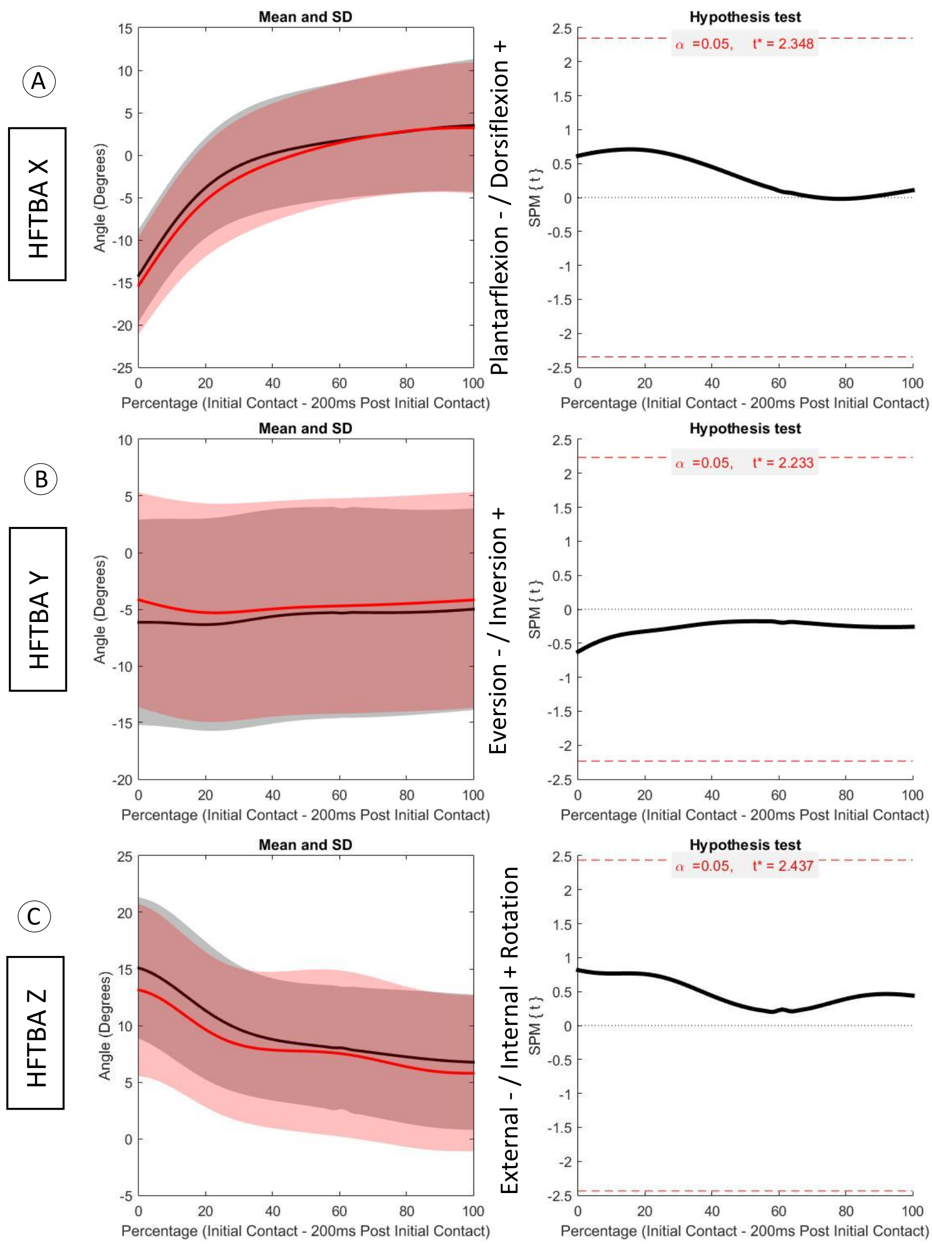


Figure H.10 Single leg landing - Hindfoot –Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

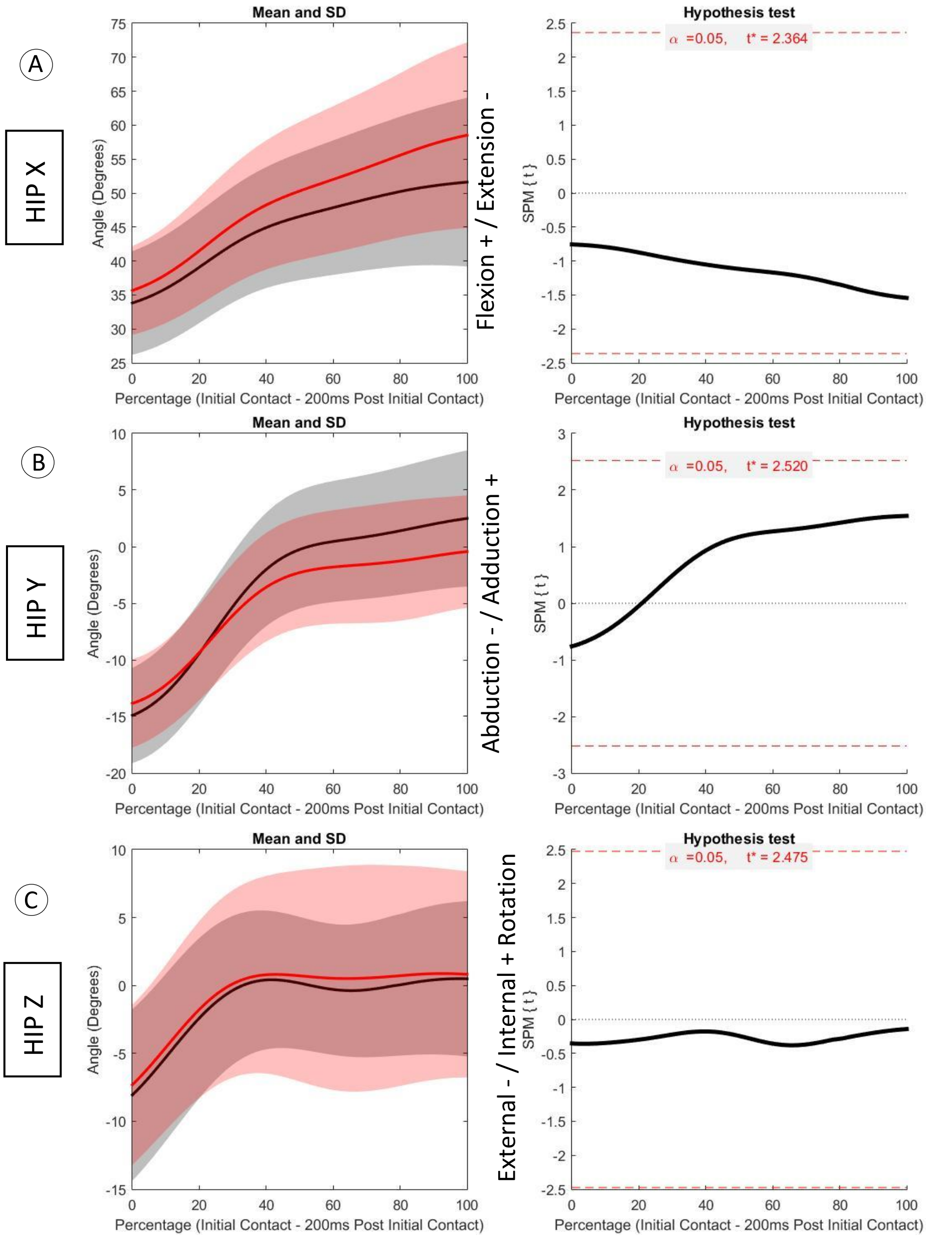


Figure H.11 Single leg landing - Hip angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

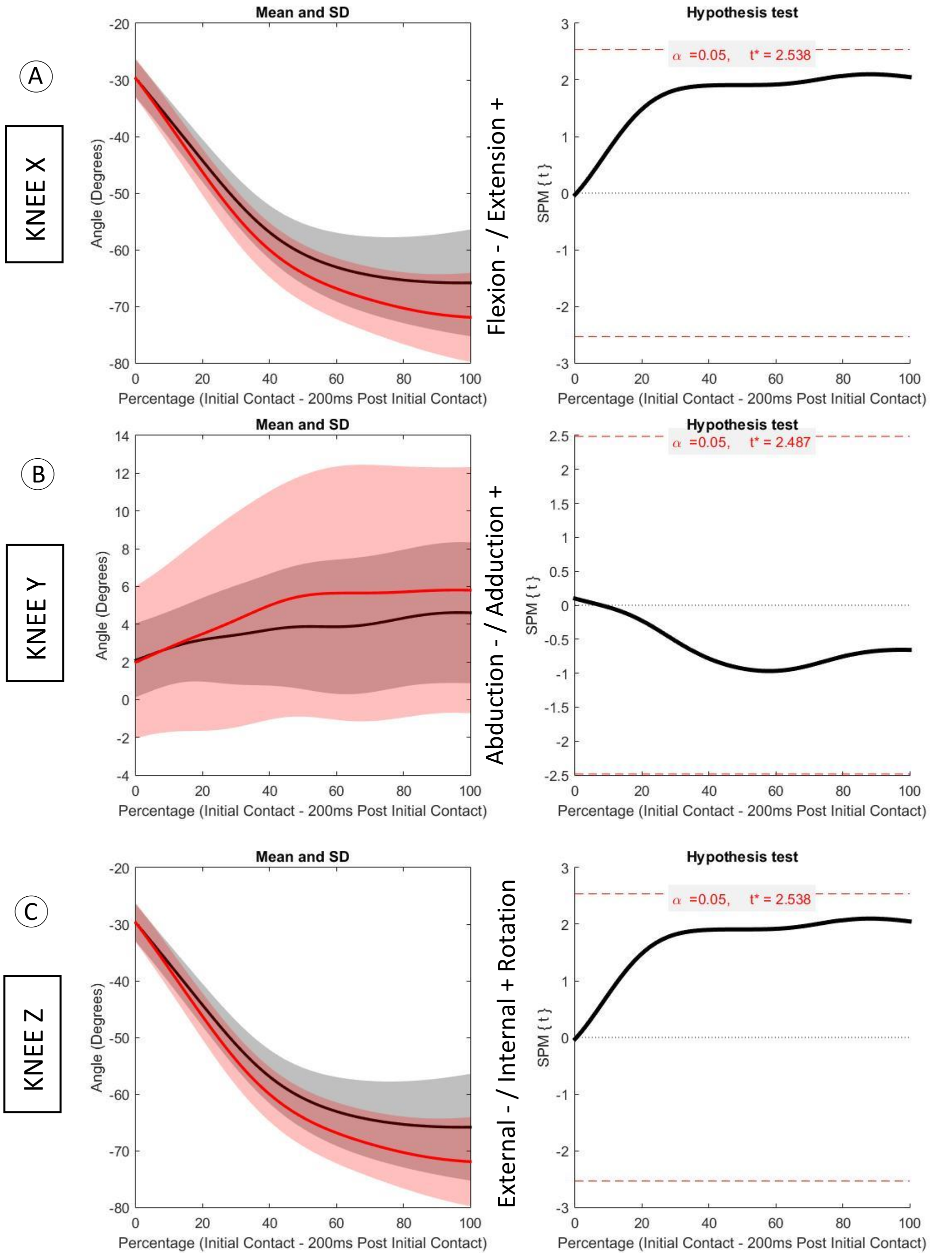


Figure H.12 Single leg landing - Knee angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

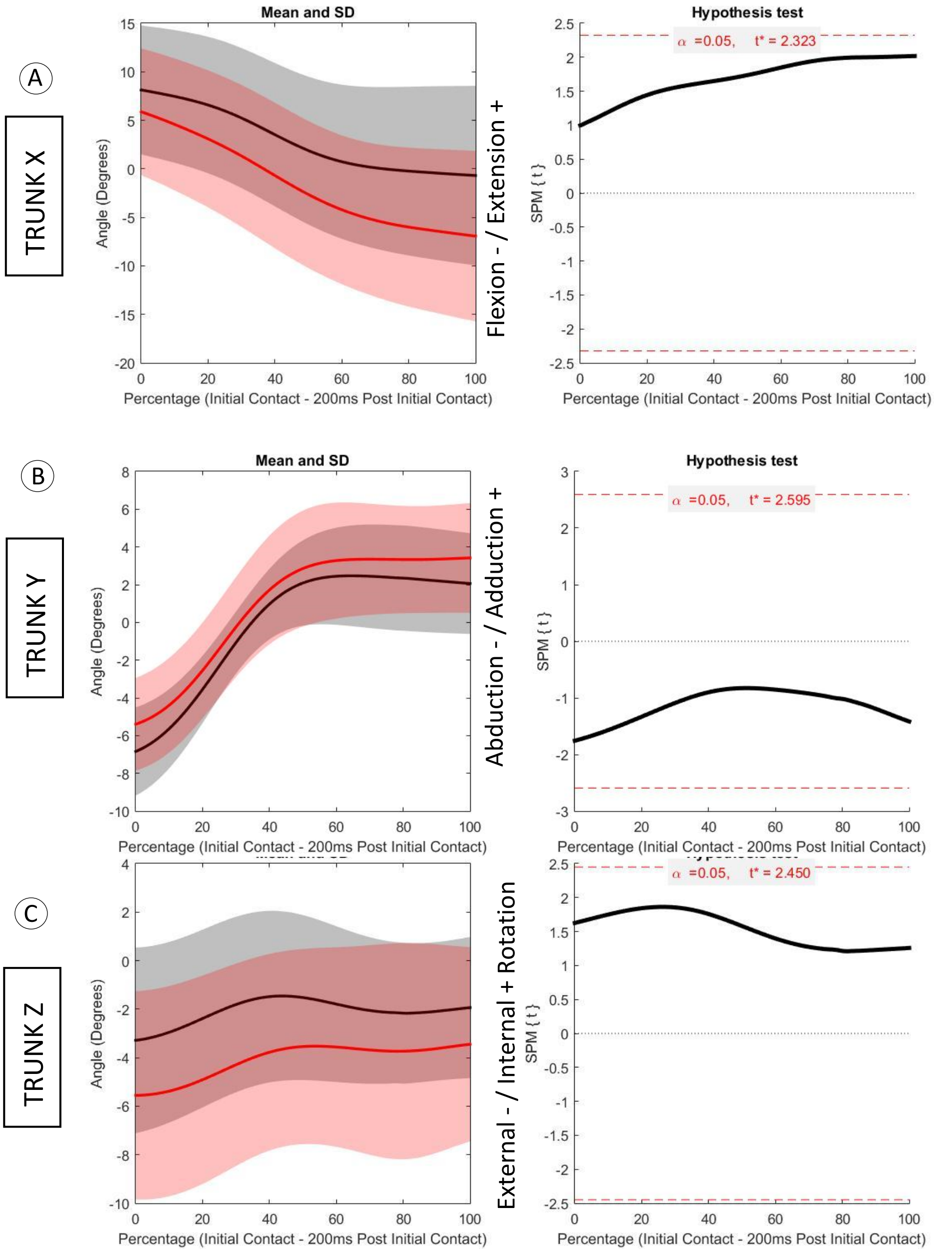


Figure H.13 Single leg landing - Trunk angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

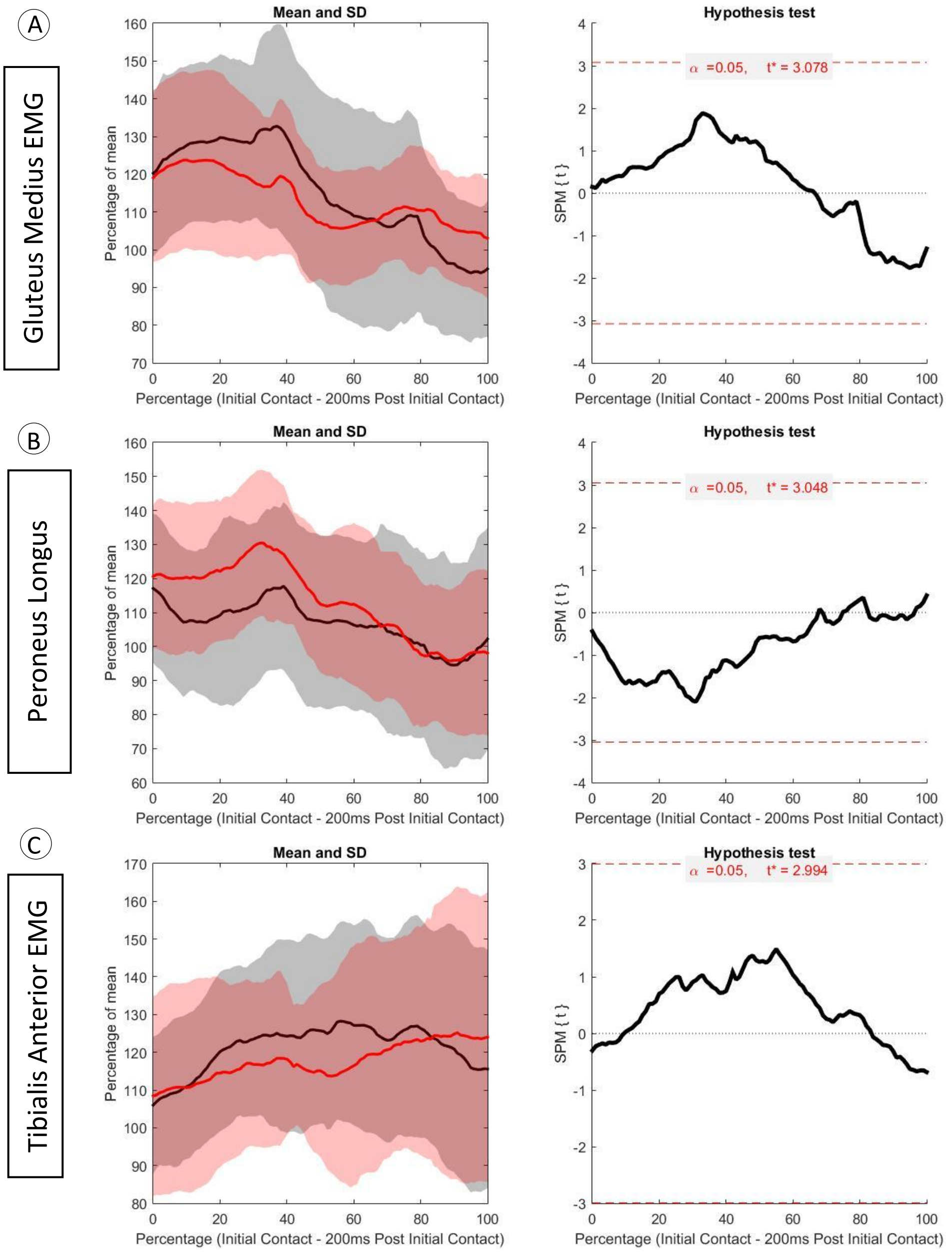


Figure H.14 Single leg landing - Gluteus Medius, Peroneus Longus and Tibialis Anterior muscle activation - IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

Unaffected limb vs Affected limb of CAI group

- **200 ms pre initial contact to initial contact**

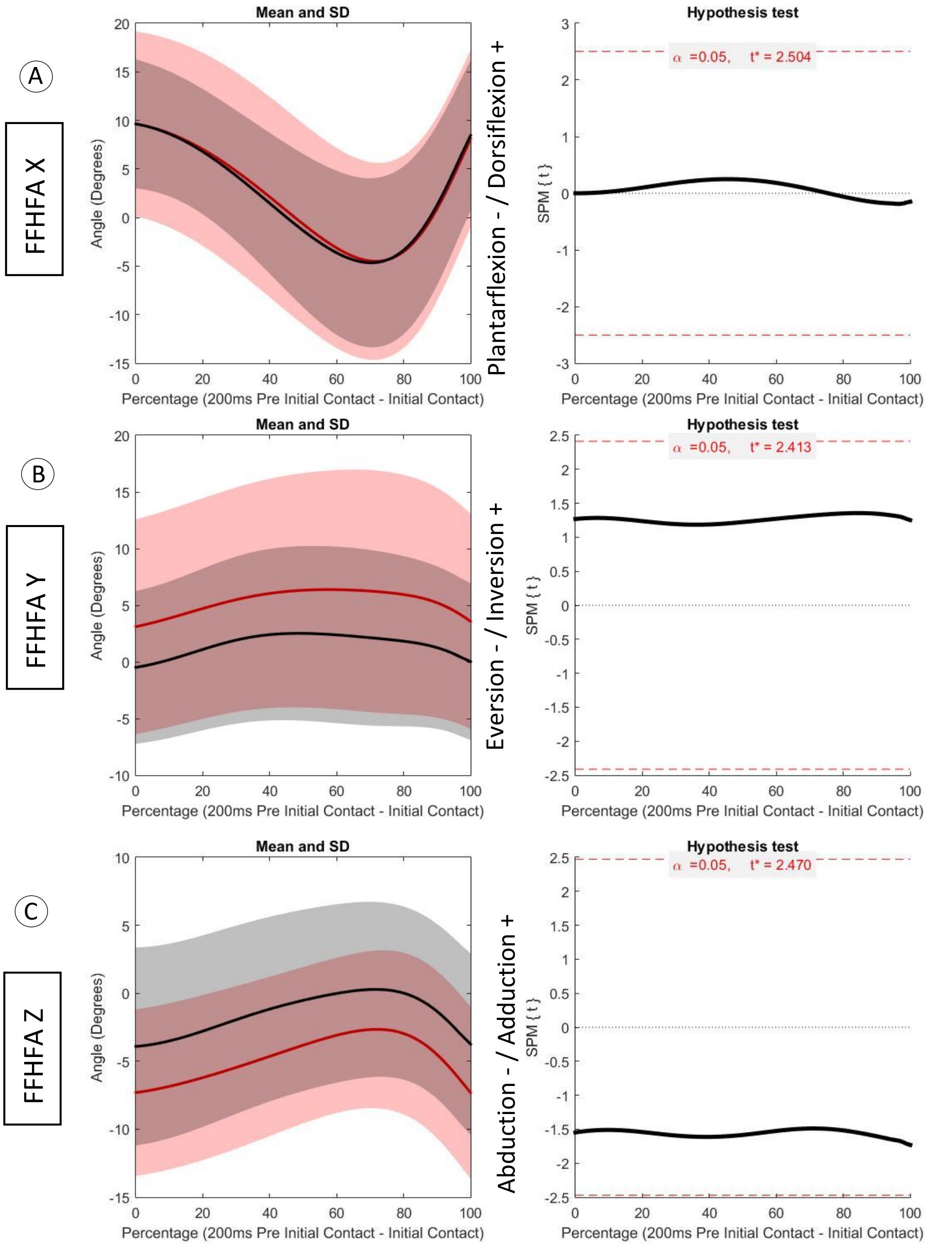


Figure H.15 Single leg landing - Forefoot – Hindfoot angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

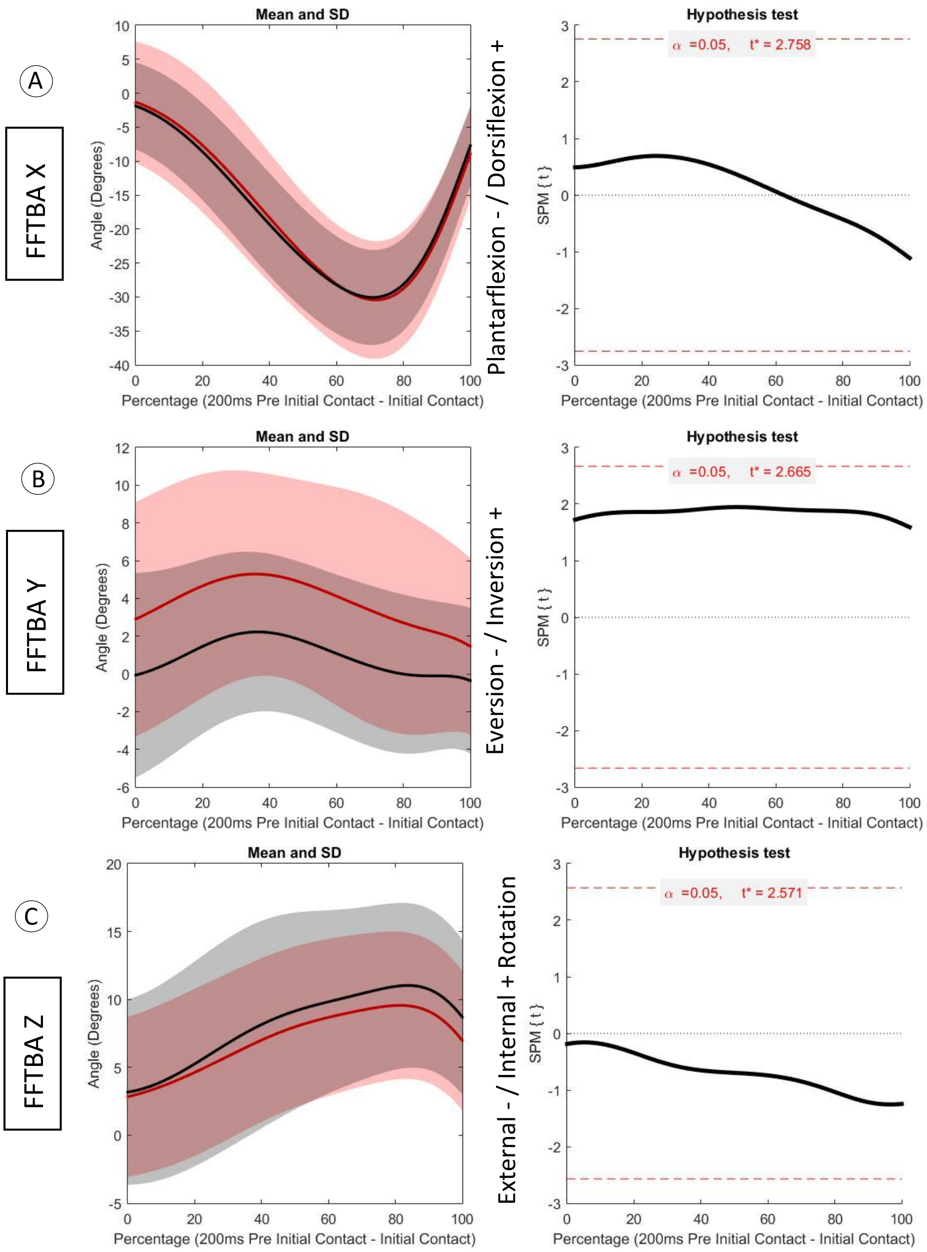


Figure H.16 Single leg landing - Forefoot – Tibia angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

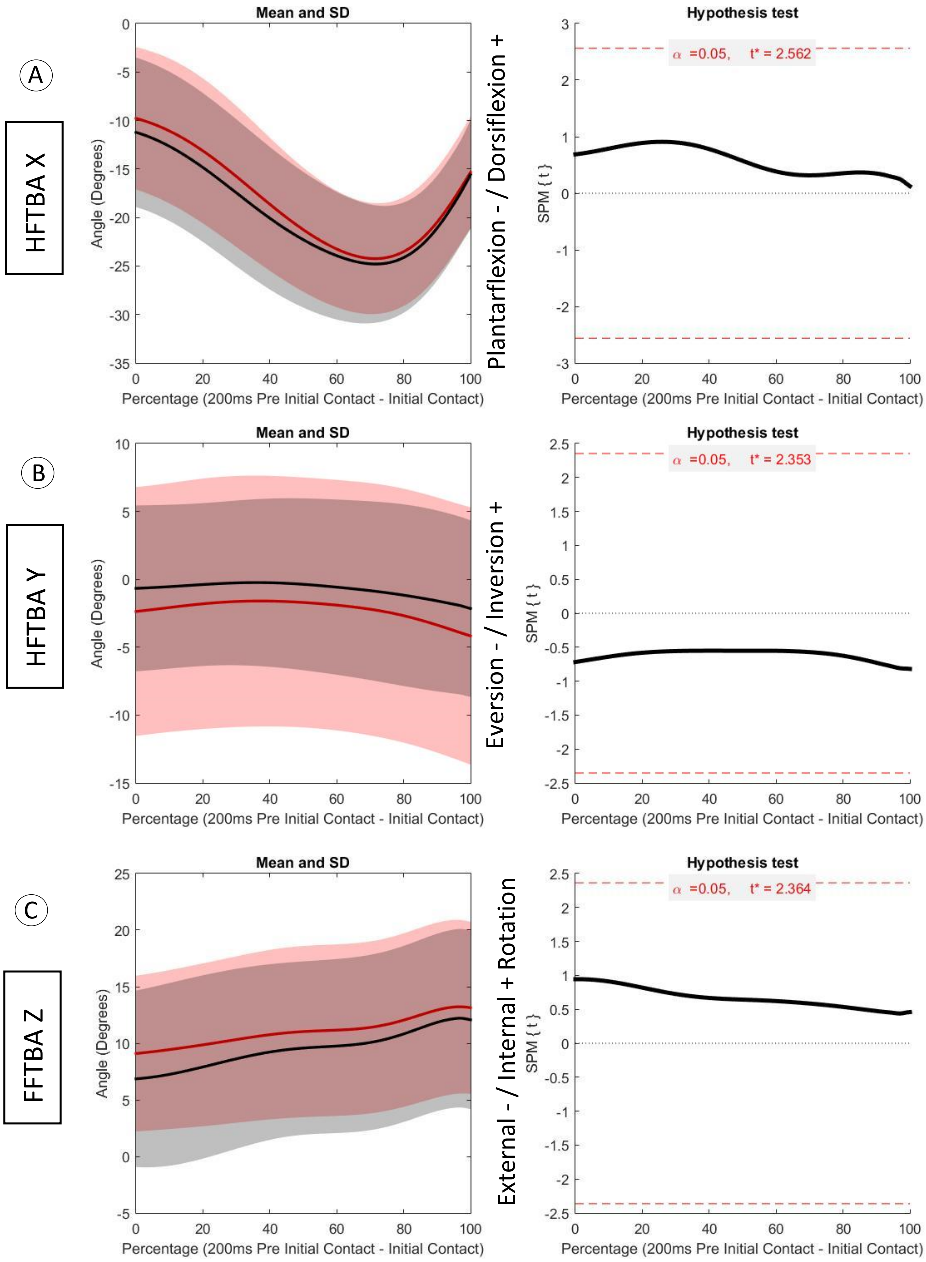


Figure H.17 Single leg landing - Hindfoot – Tibia angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

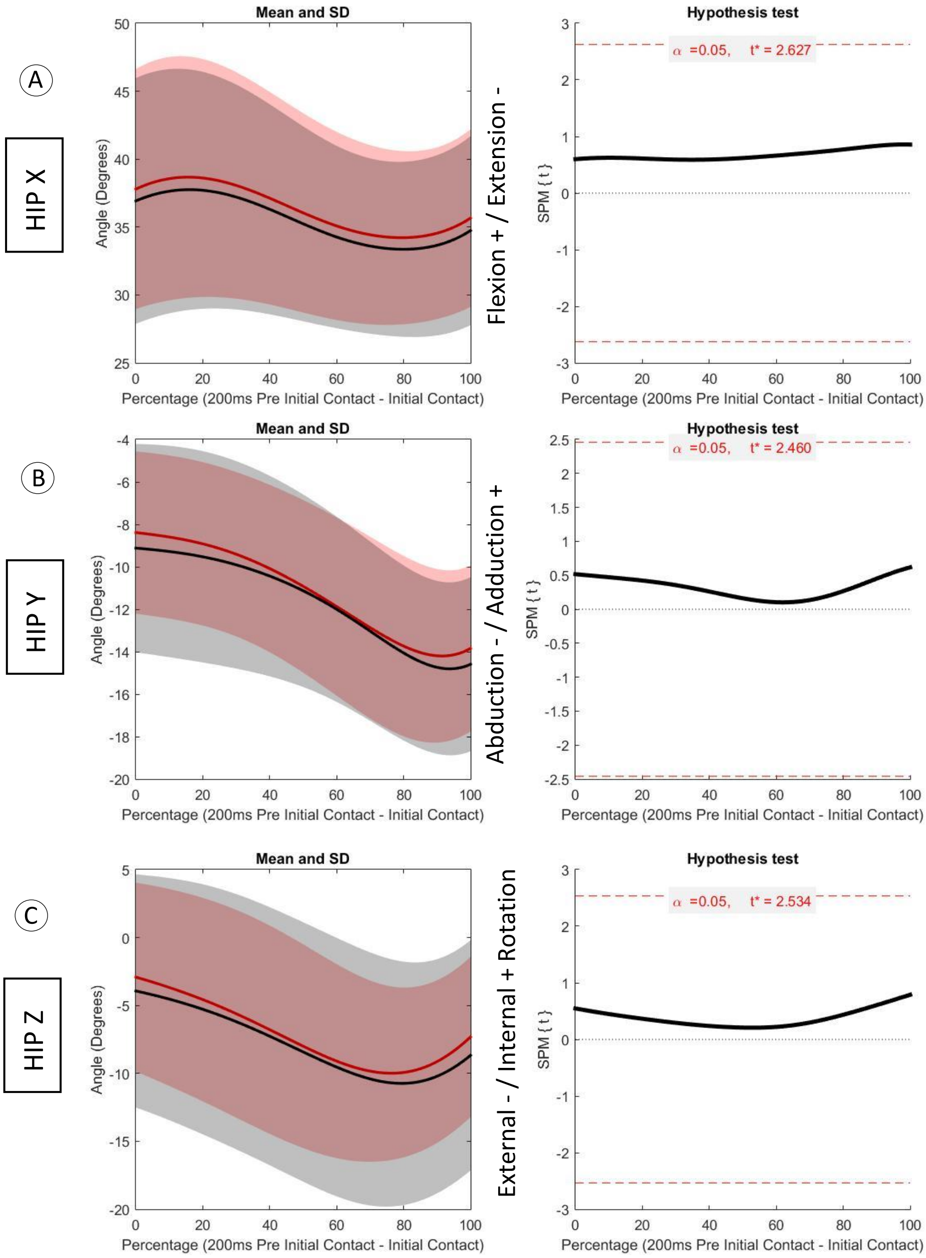


Figure H.18 Single leg landing - Hip angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

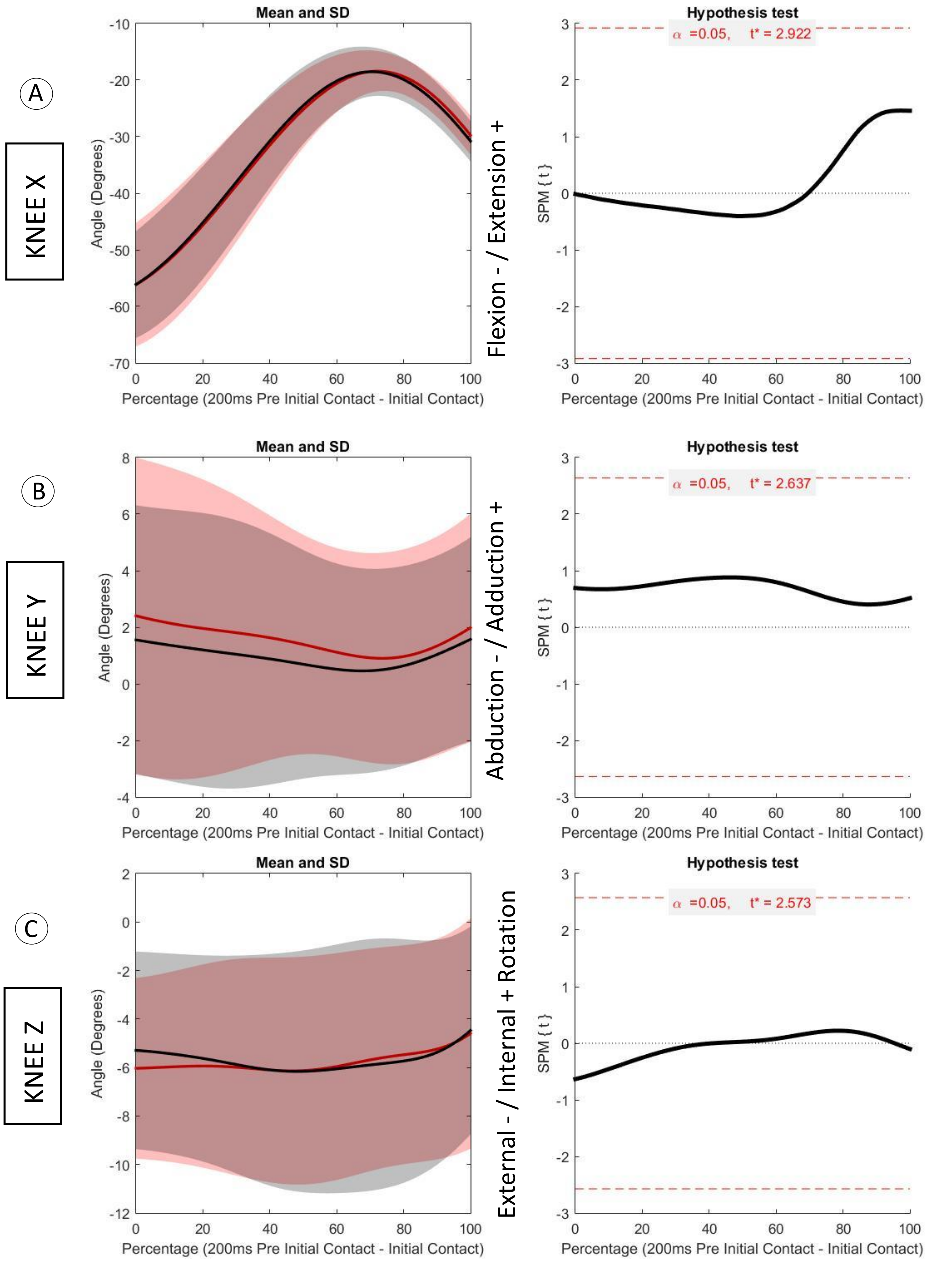


Figure H.19 Single leg landing - Knee angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

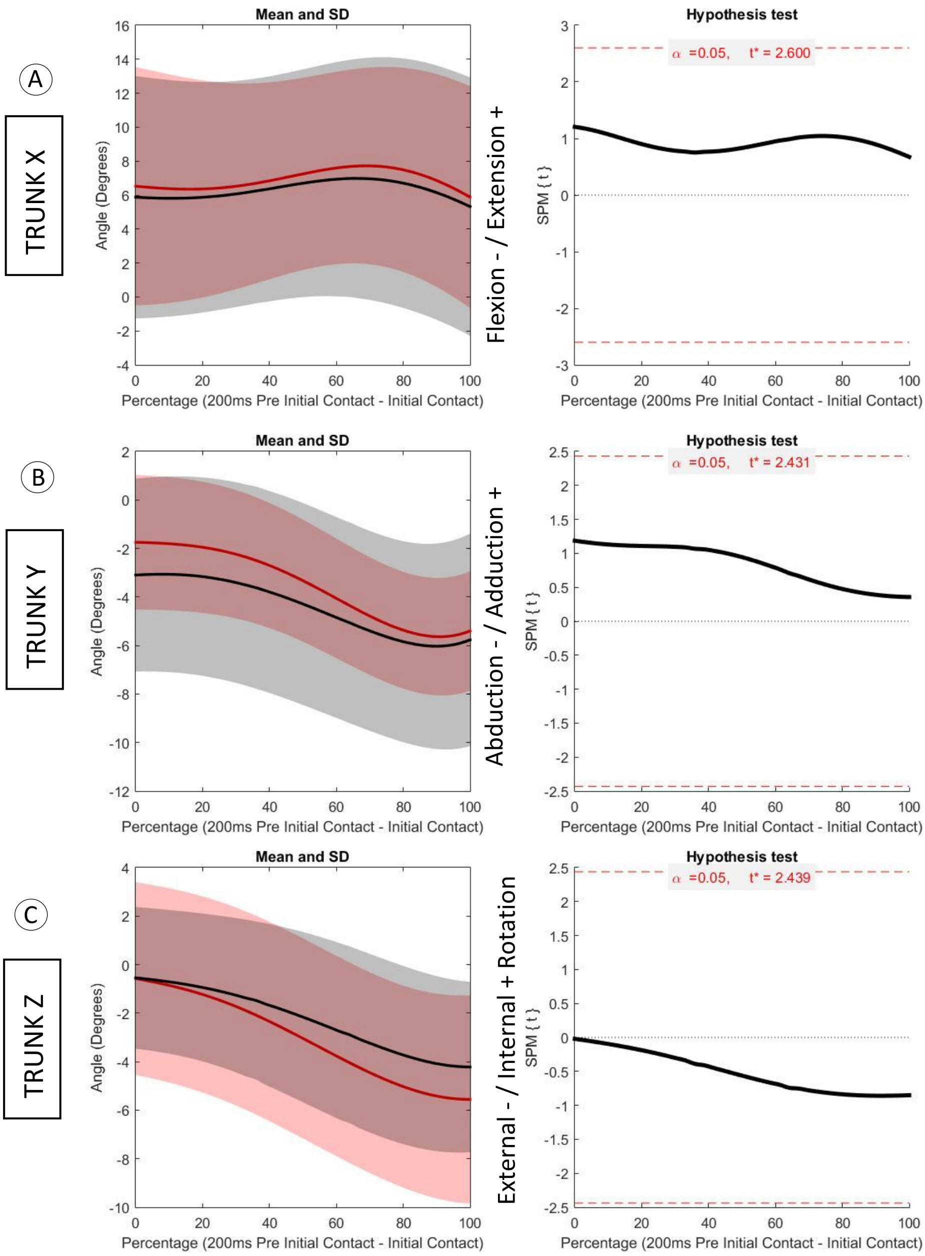


Figure H.20 Single leg landing - Trunk angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

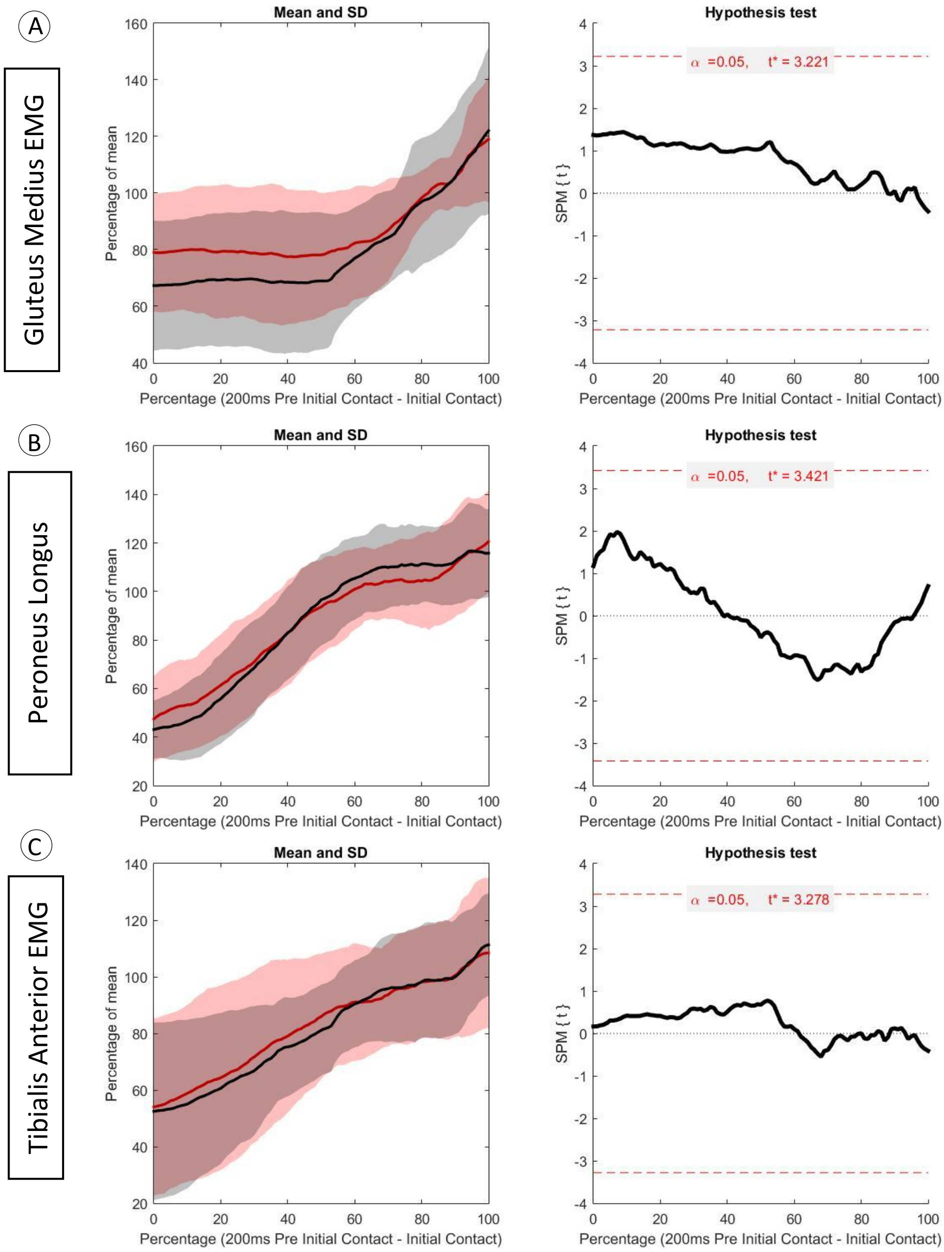


Figure H.21 Single leg landing - Gluteus Medius, Peroneus Longus and Tibialis Anterior muscle activation - 200 ms pre IC to IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

- **Initial contact to 200 ms post initial contact**

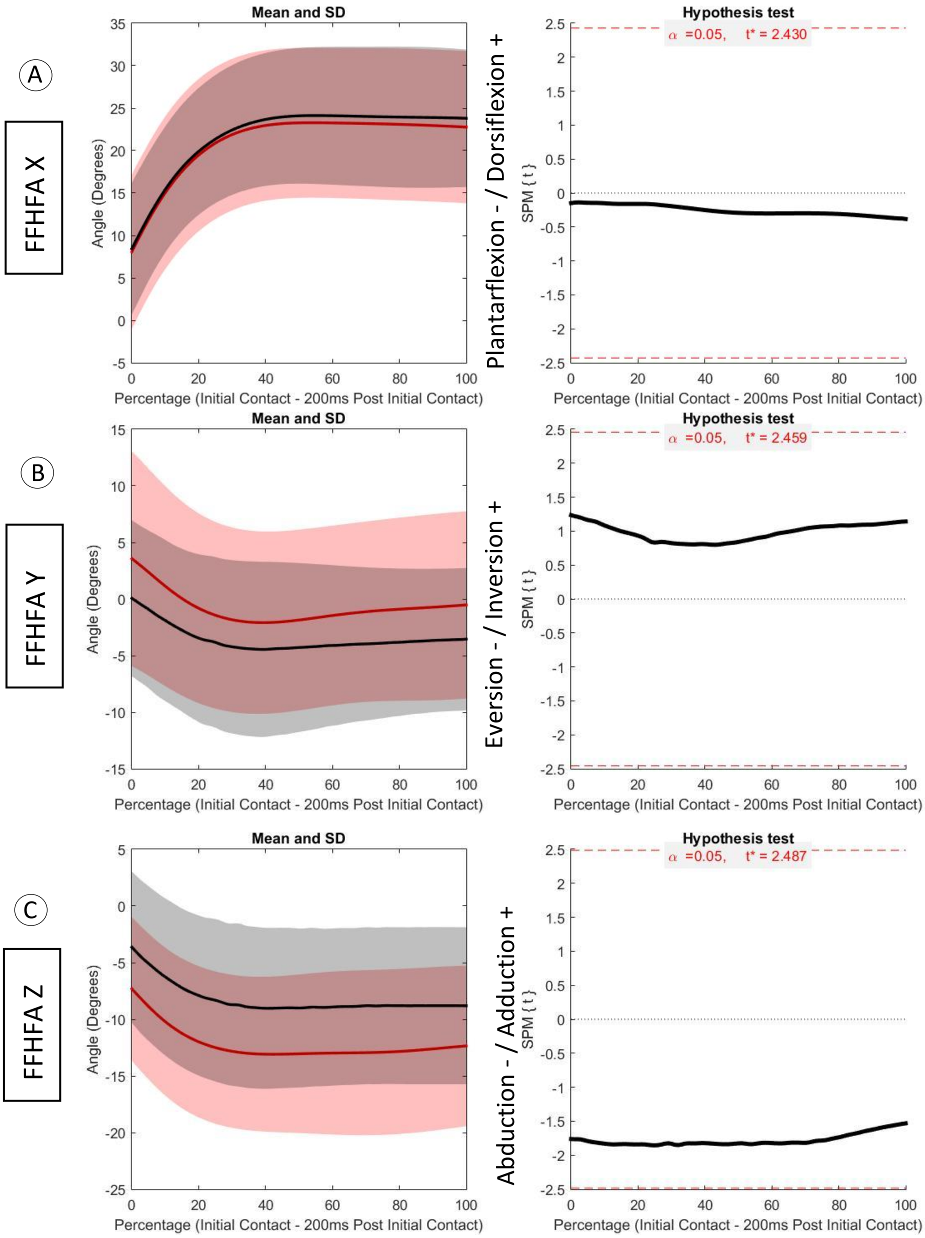


Figure H.22 Single leg landing - Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

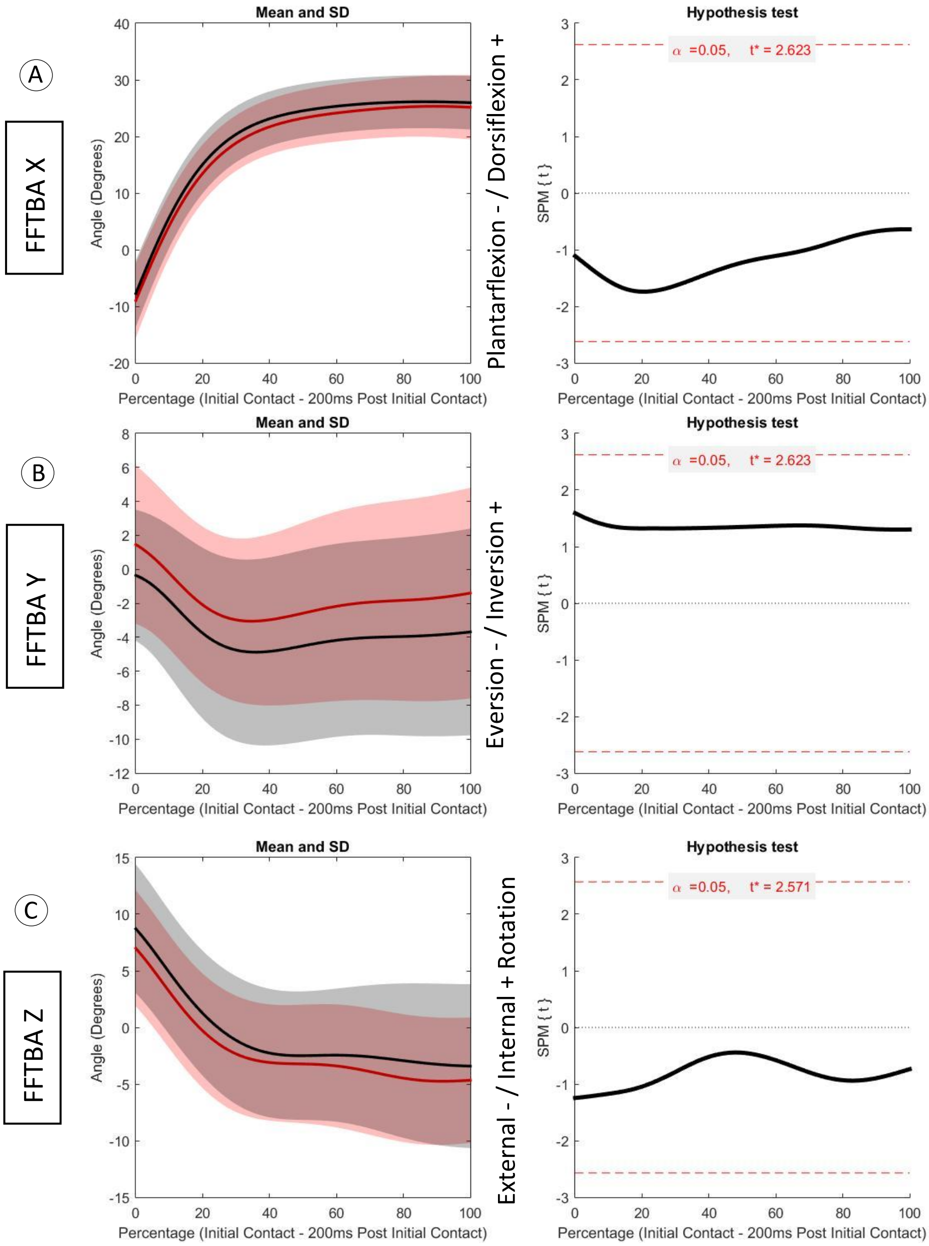


Figure H.23 Single leg landing - Forefoot – Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

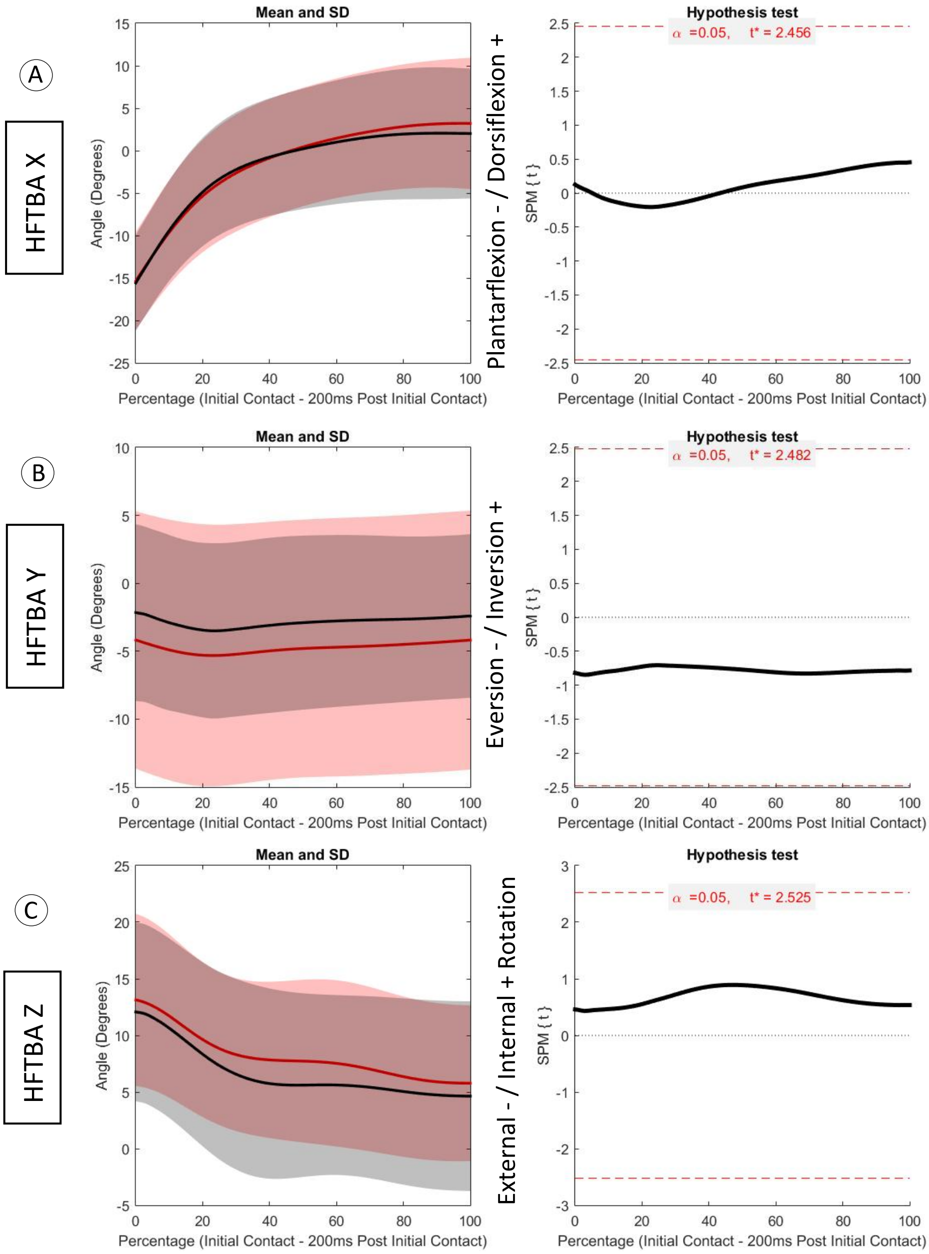


Figure H.24 Single leg landing - Hindfoot – Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

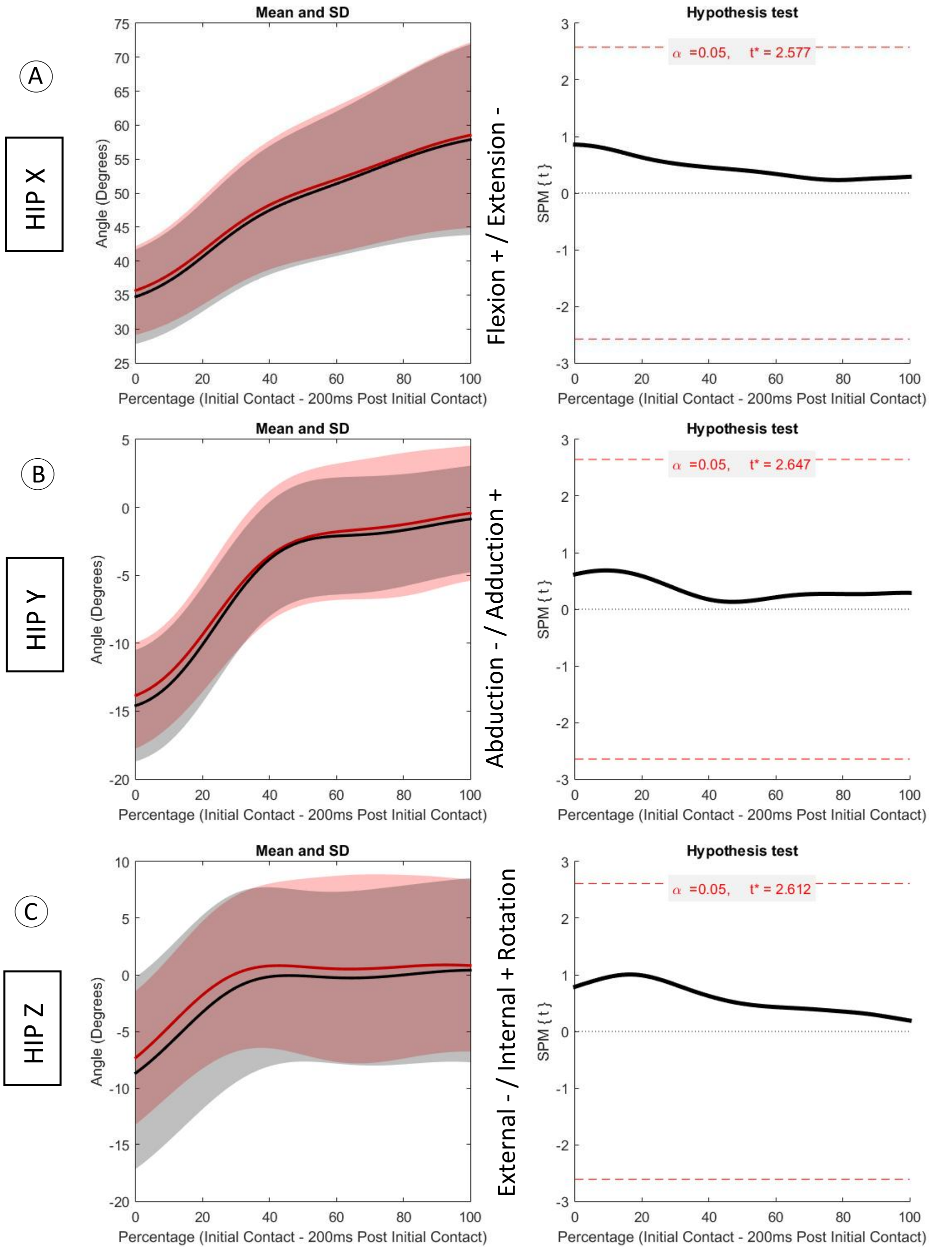


Figure H.25 Single leg landing - Hip angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

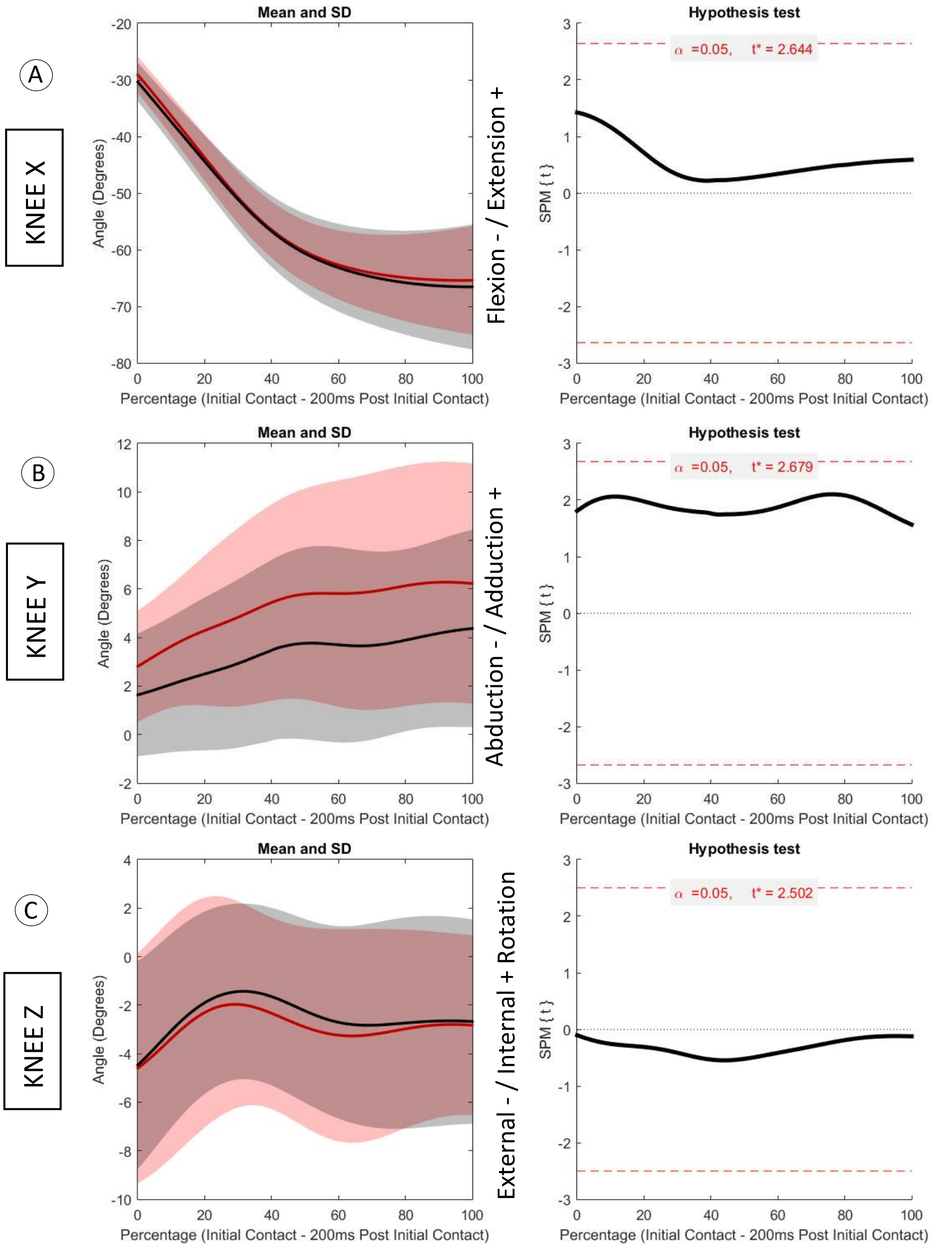


Figure H.26 Single leg landing - Knee angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

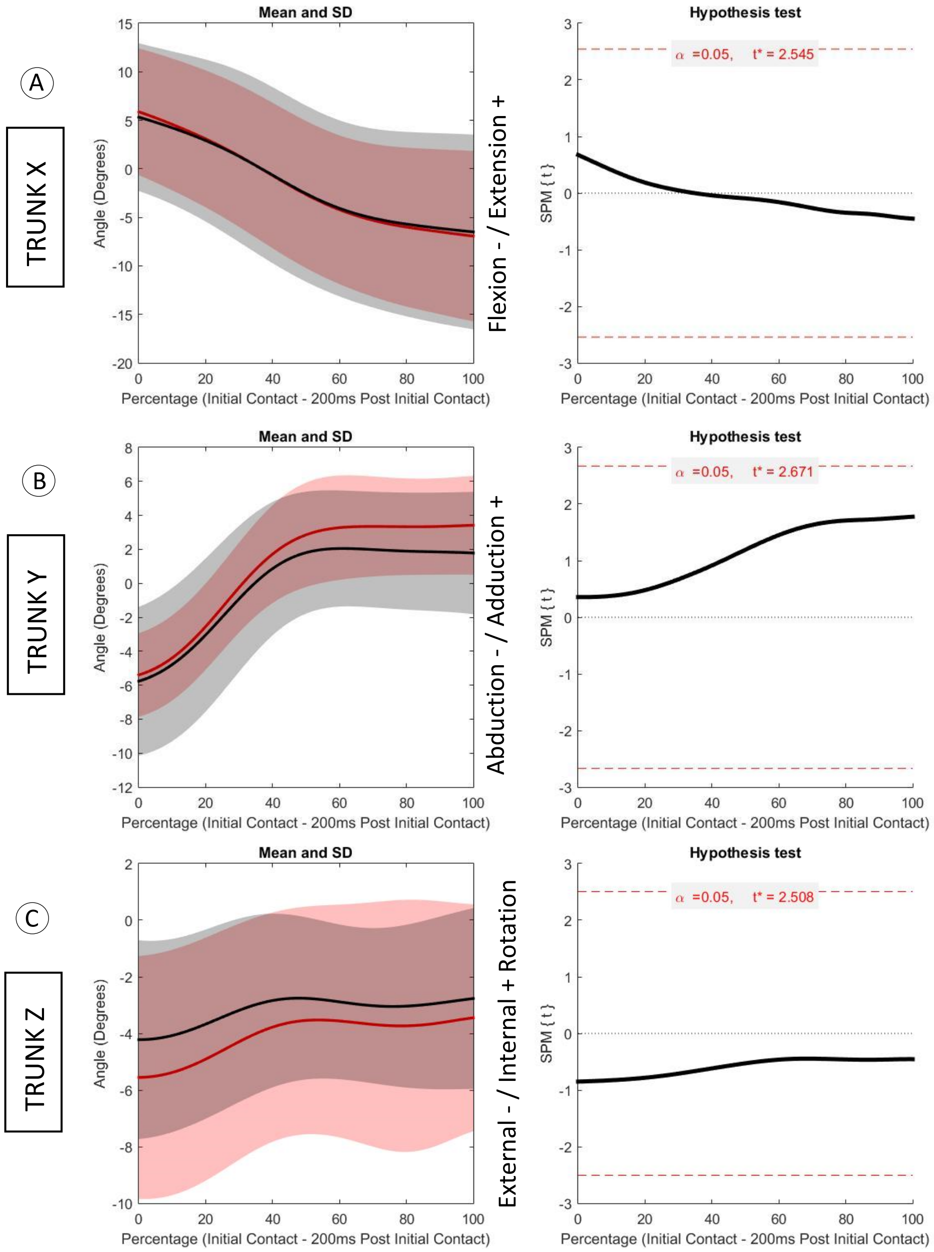


Figure H.27 Single leg landing - Trunk angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

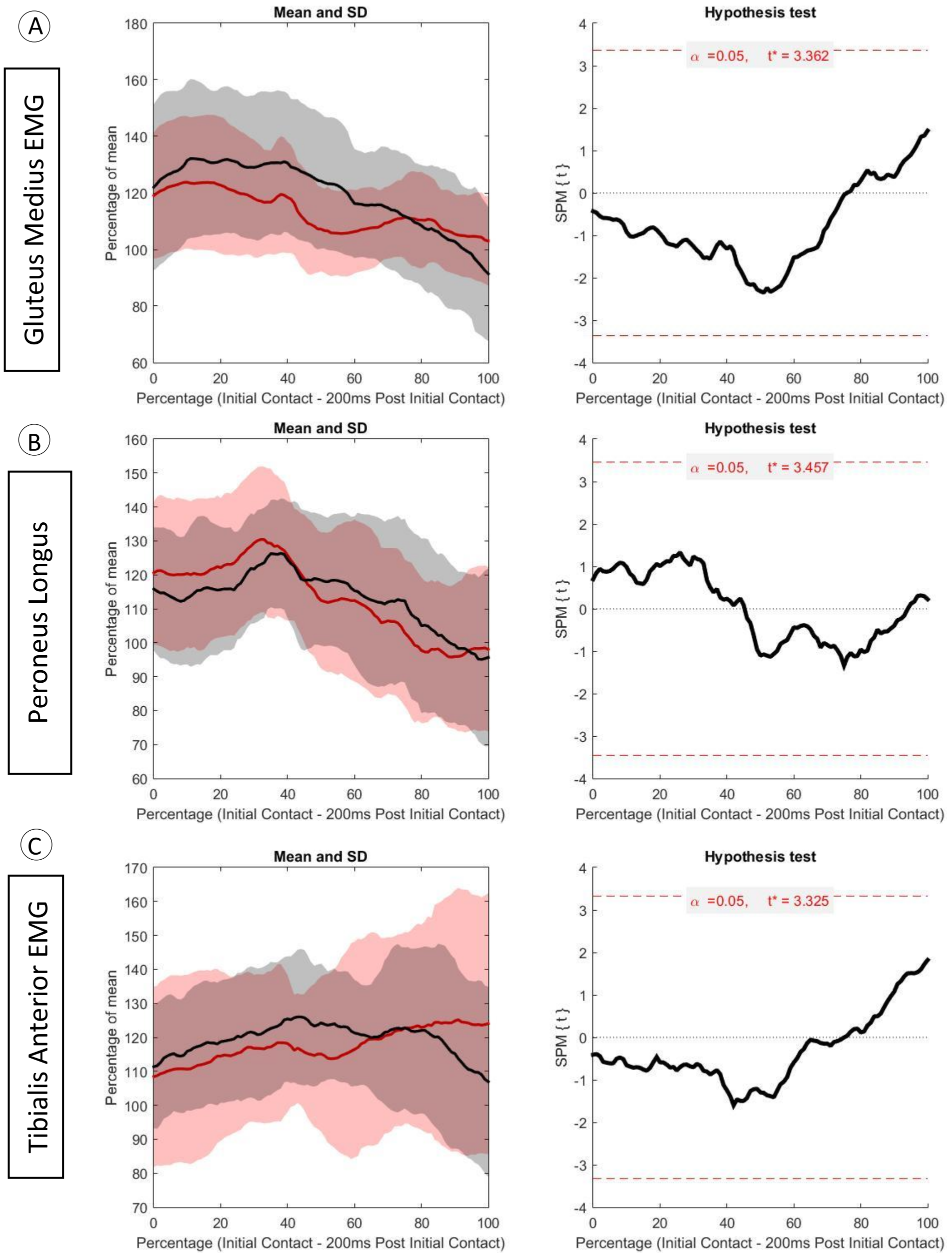


Figure H.28 Single leg landing - Gluteus Medius, Peroneus Longus and Tibialis Anterior muscle activation - IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

Matched Control vs Unaffected limb of CAI group

- **200 ms pre initial contact to initial contact**

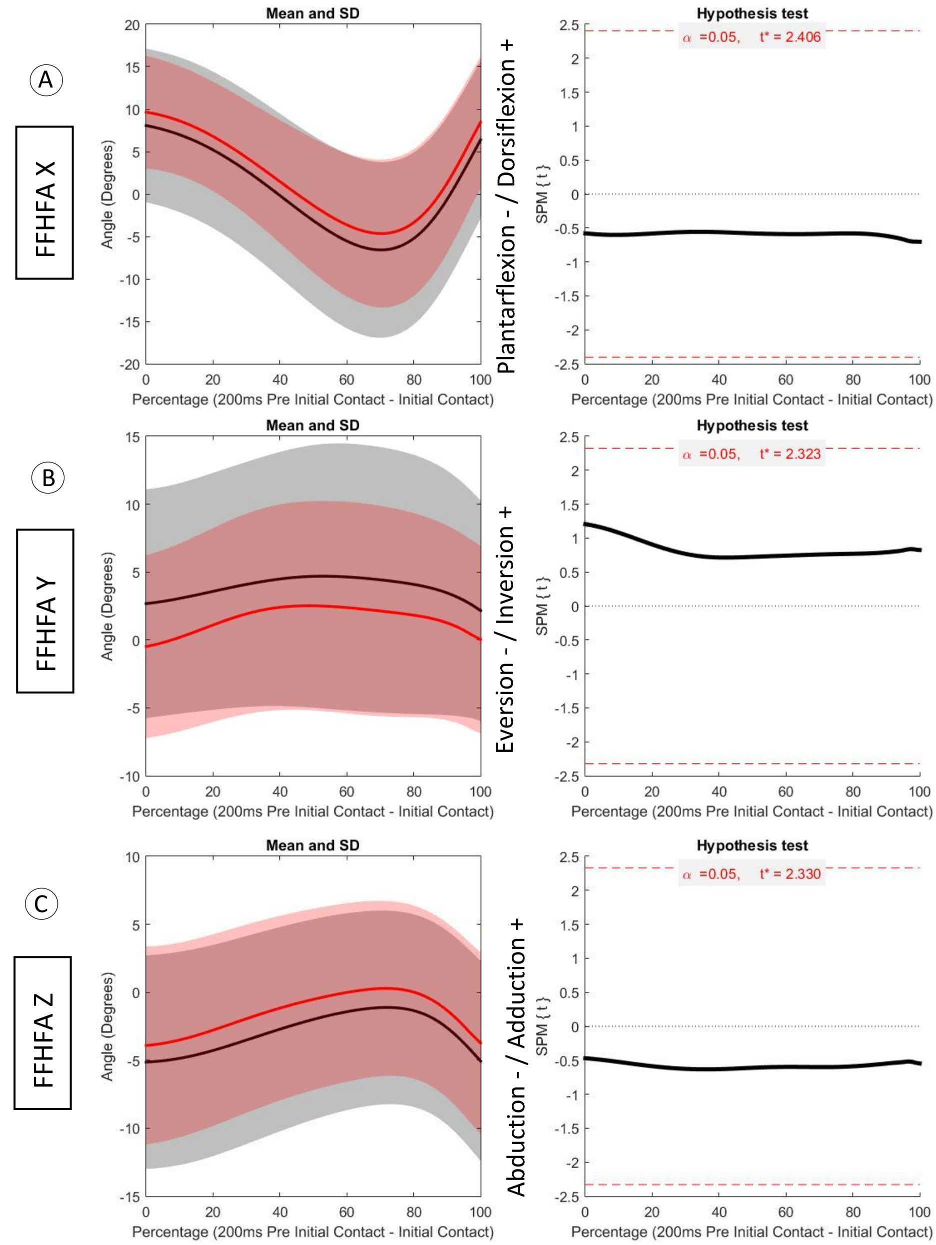


Figure H.29 Single leg landing - Forefoot – Hindfoot angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

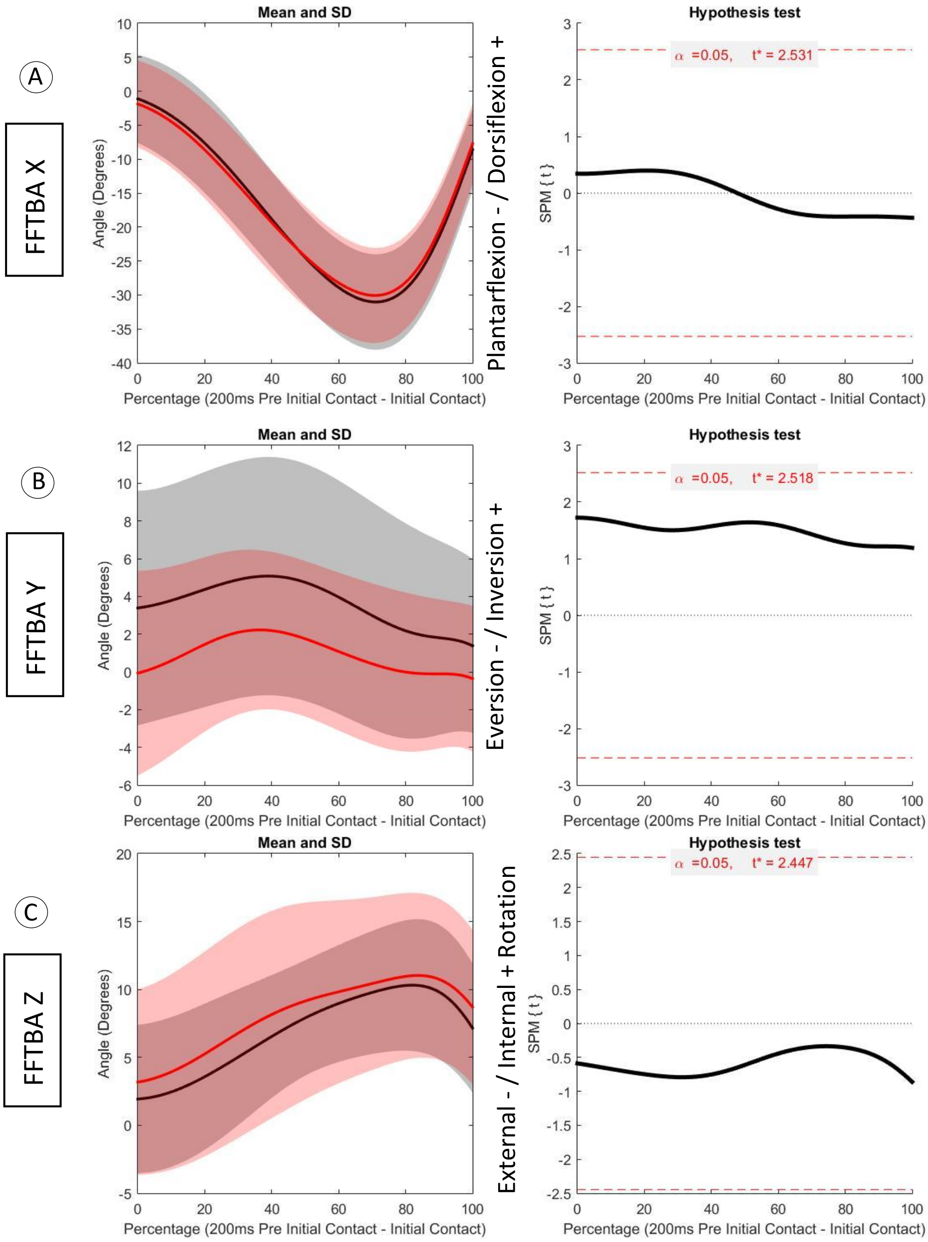


Figure H.30 Single leg landing - Forefoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

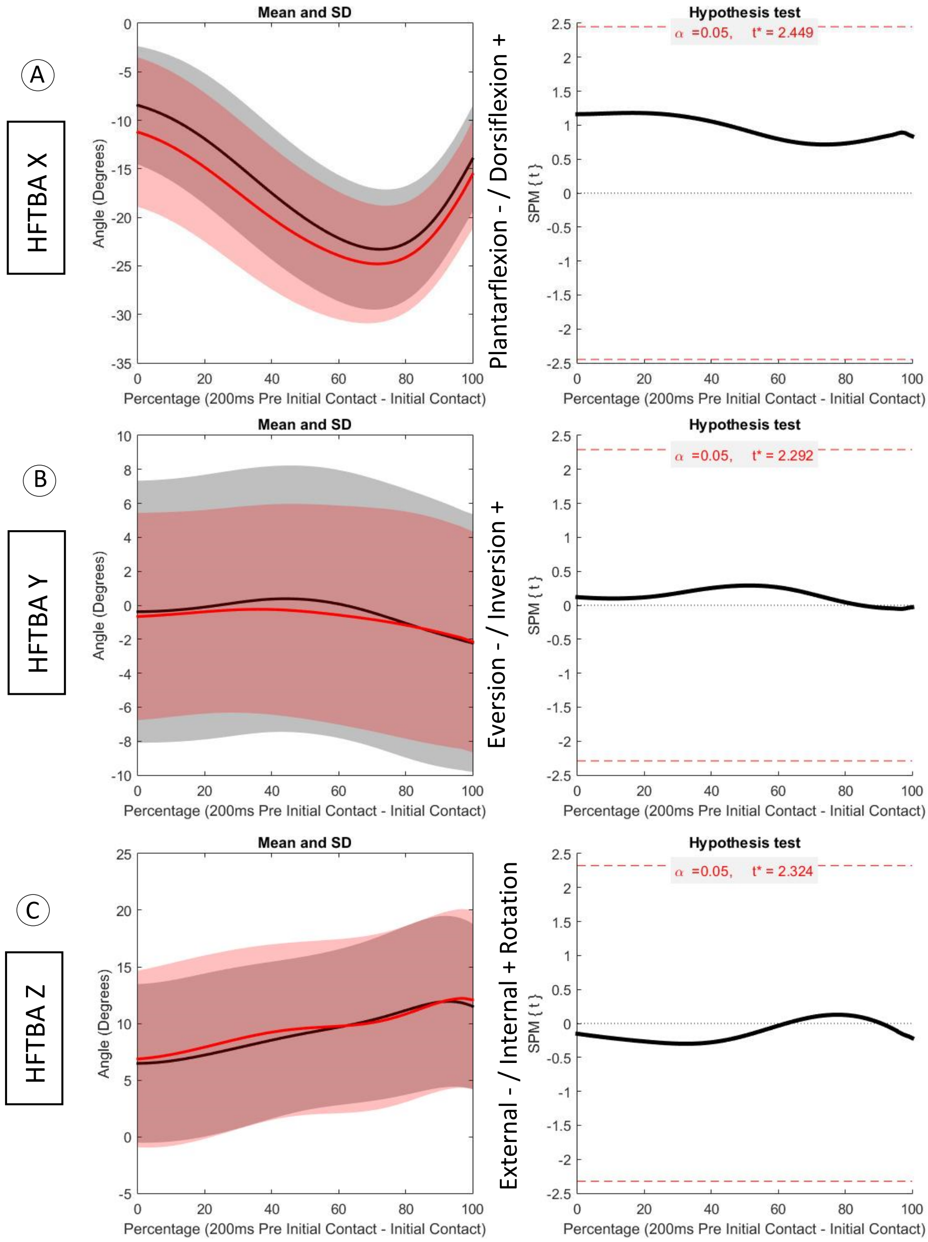


Figure H.31 Single leg landing - Hindfoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

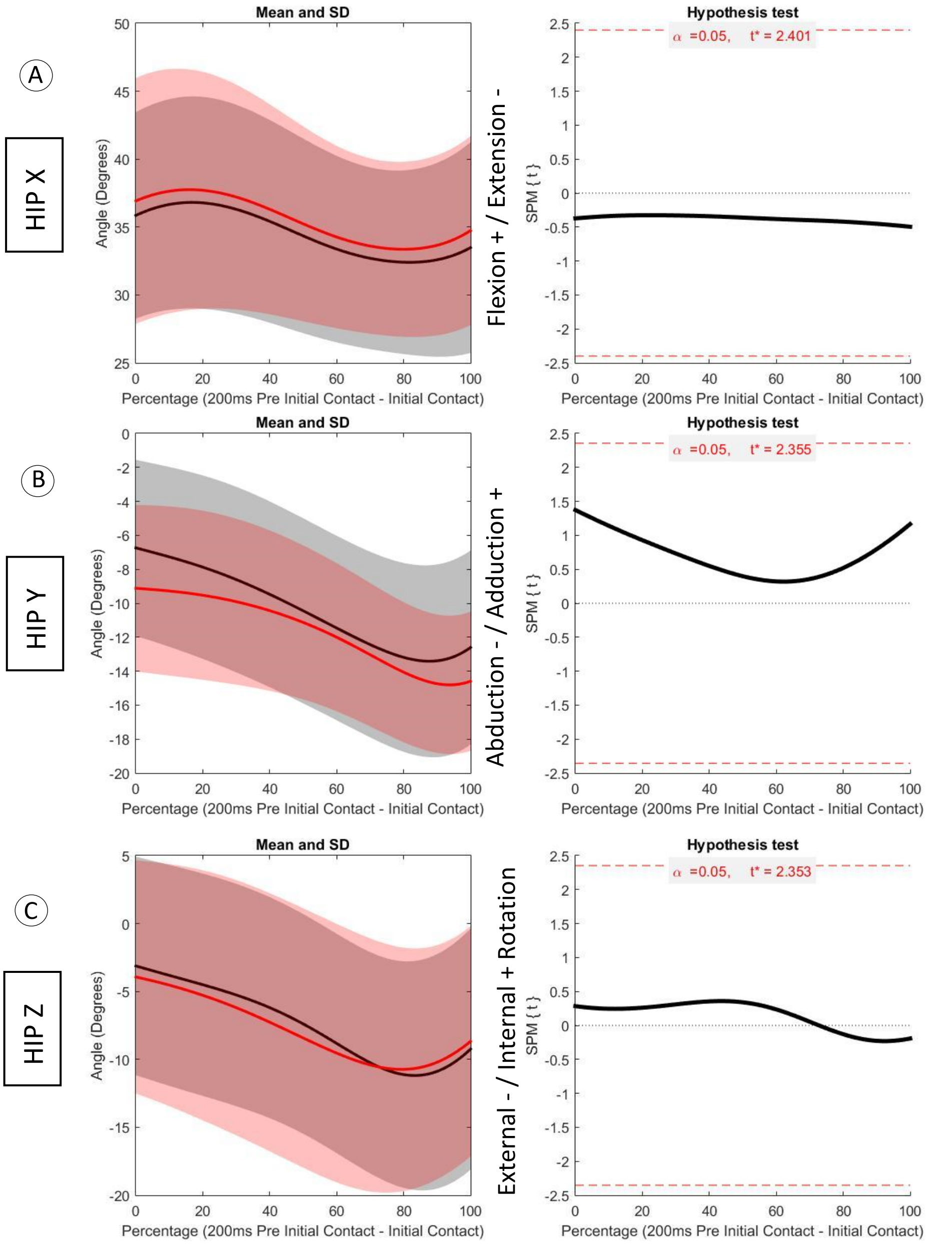


Figure H.32 Single leg landing - Hip angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

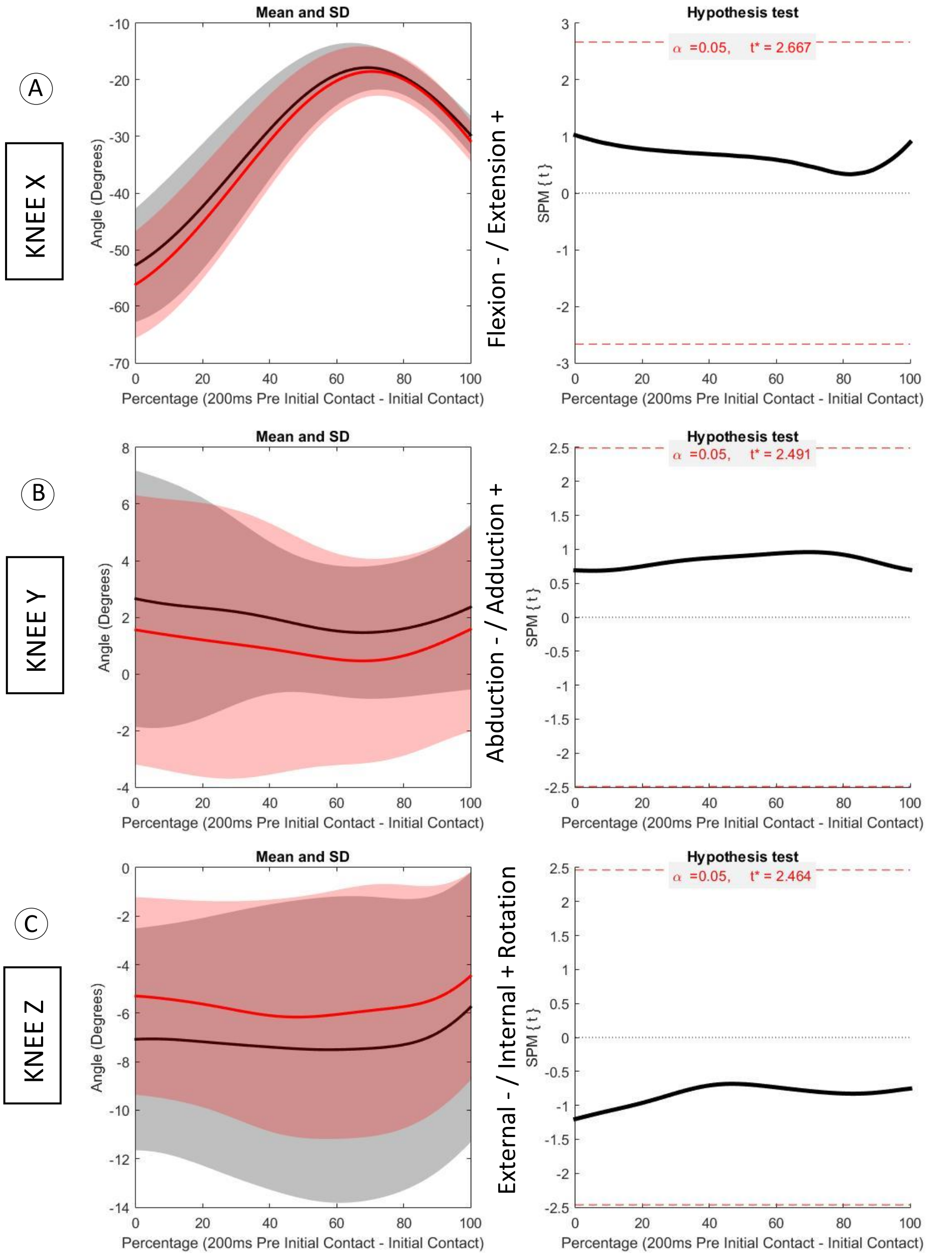


Figure H.33 Single leg landing - Knee angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

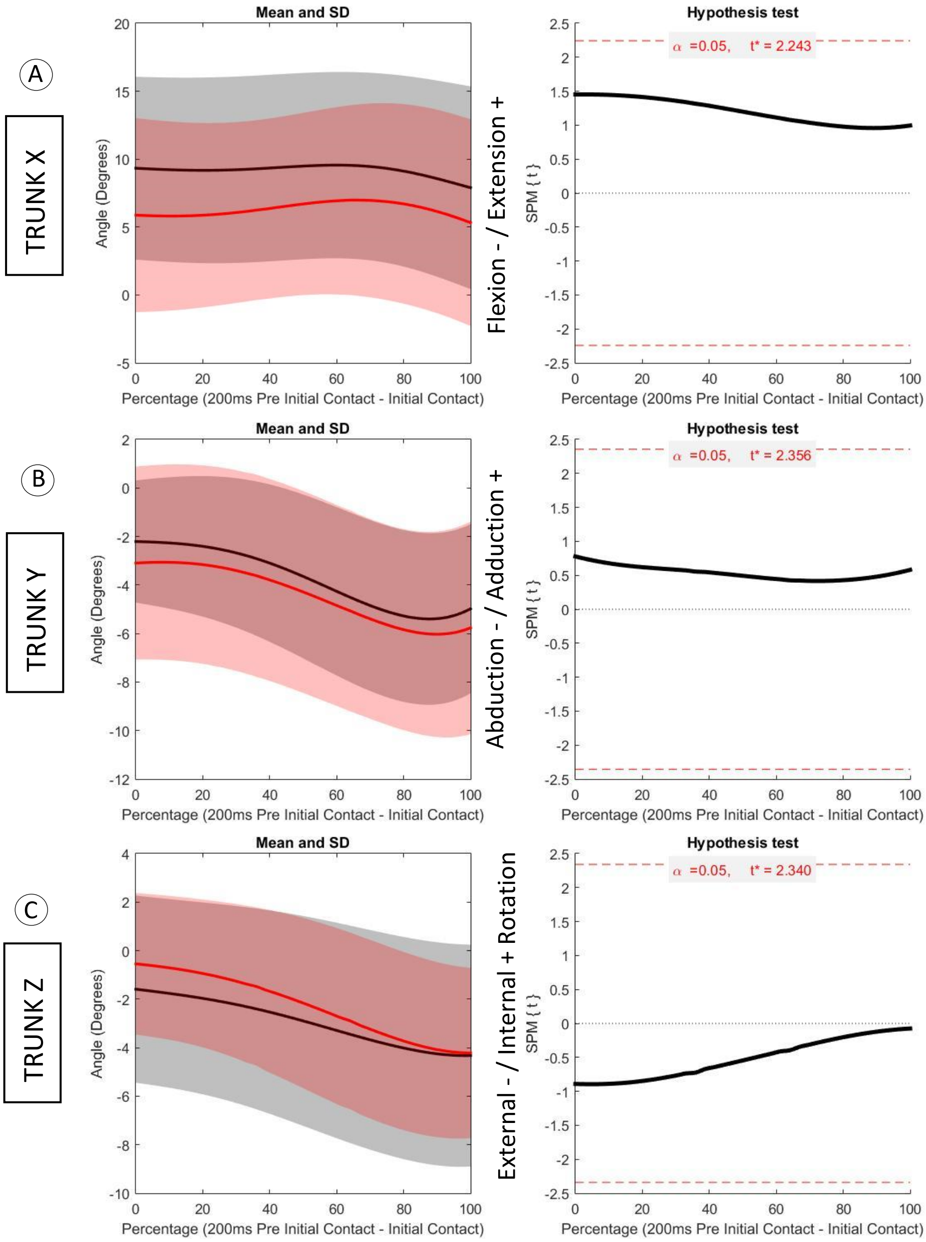


Figure H.34 Single leg landing - Trunk angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

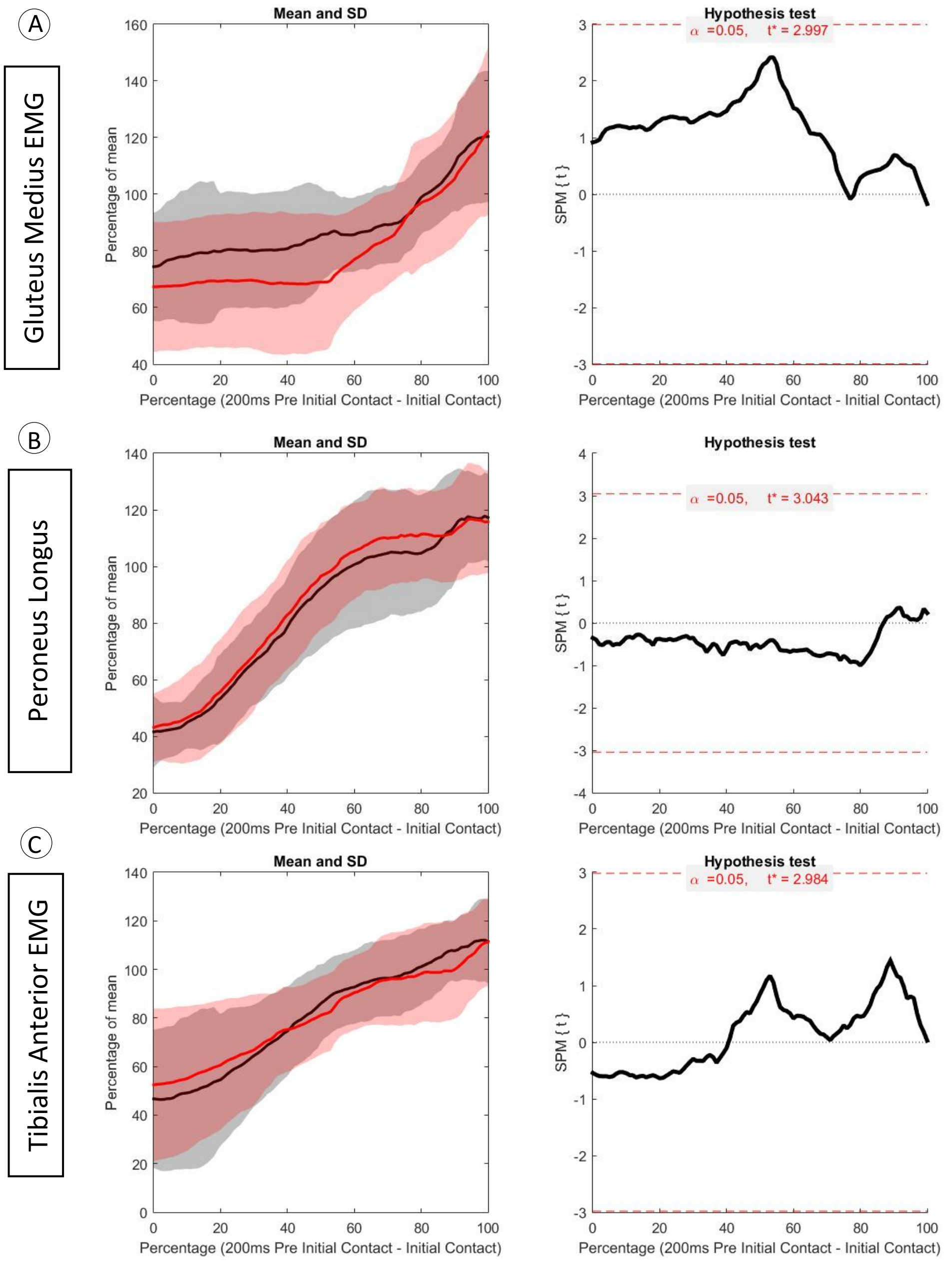


Figure H.35 Single leg landing - Gluteus Medius, Peroneus Longus and Tibialis Anterior muscle activation - 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

- **Initial contact to 200 ms post initial contact**

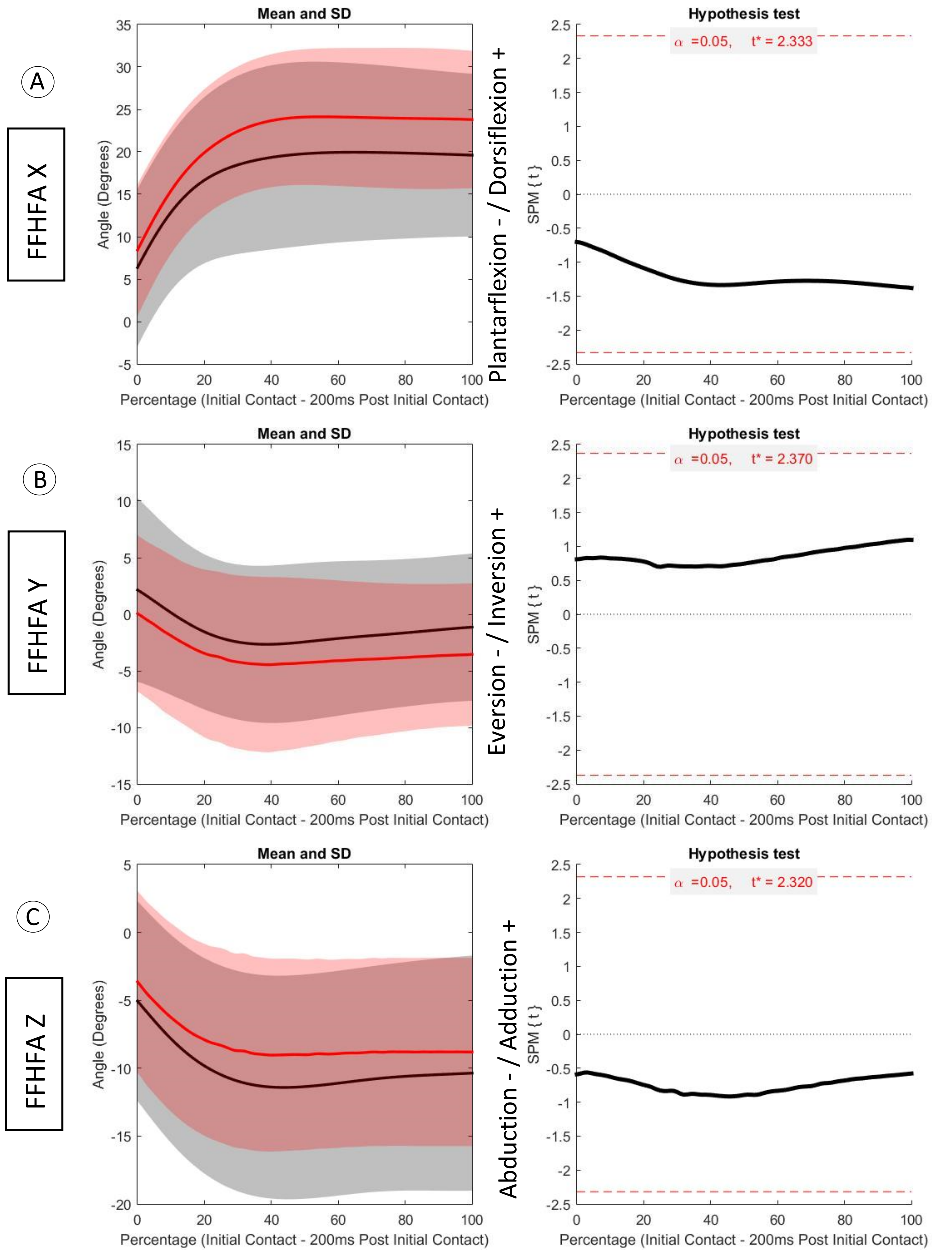


Figure H.36 Single leg landing - Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

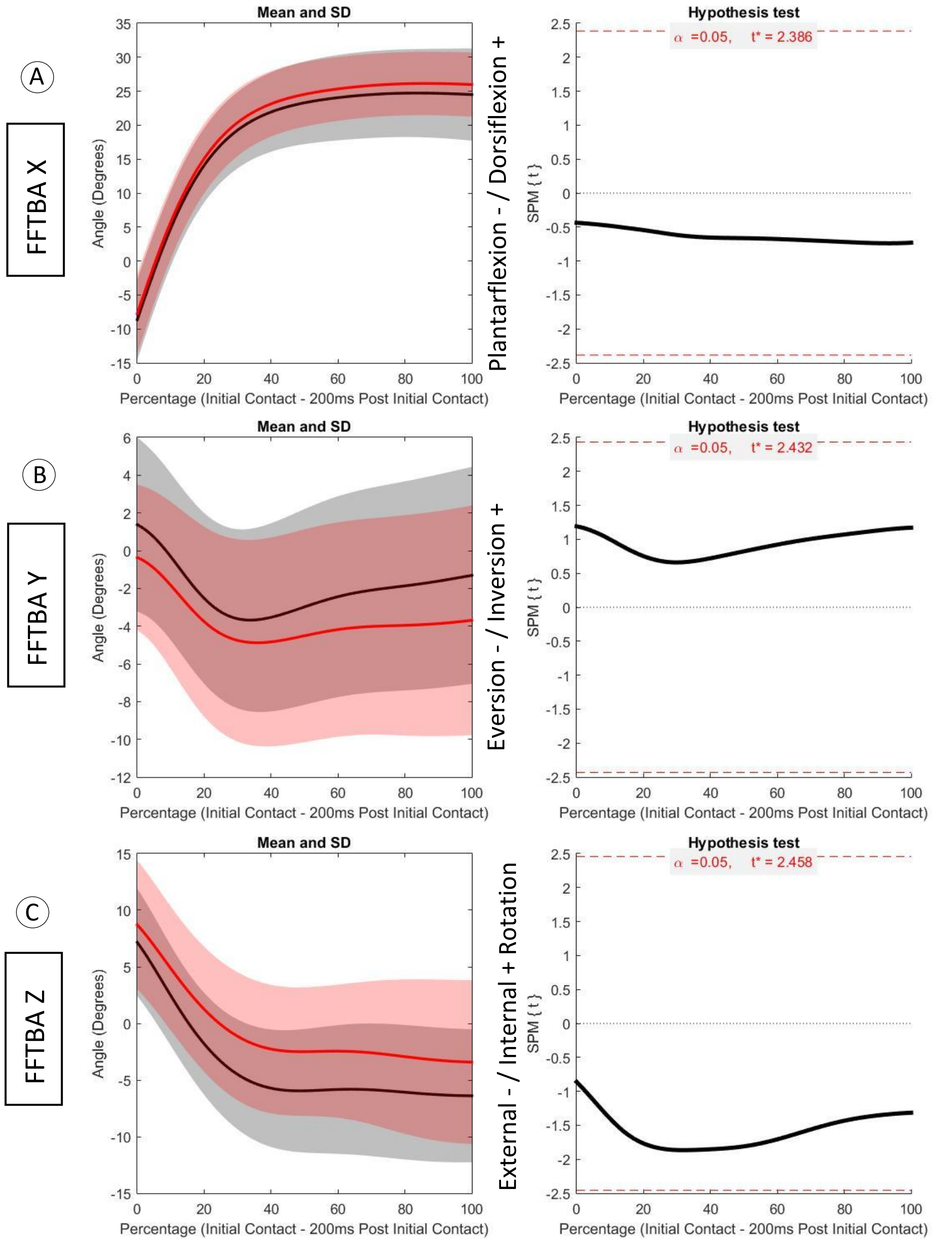


Figure H.37 Single leg landing - Forefoot –Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

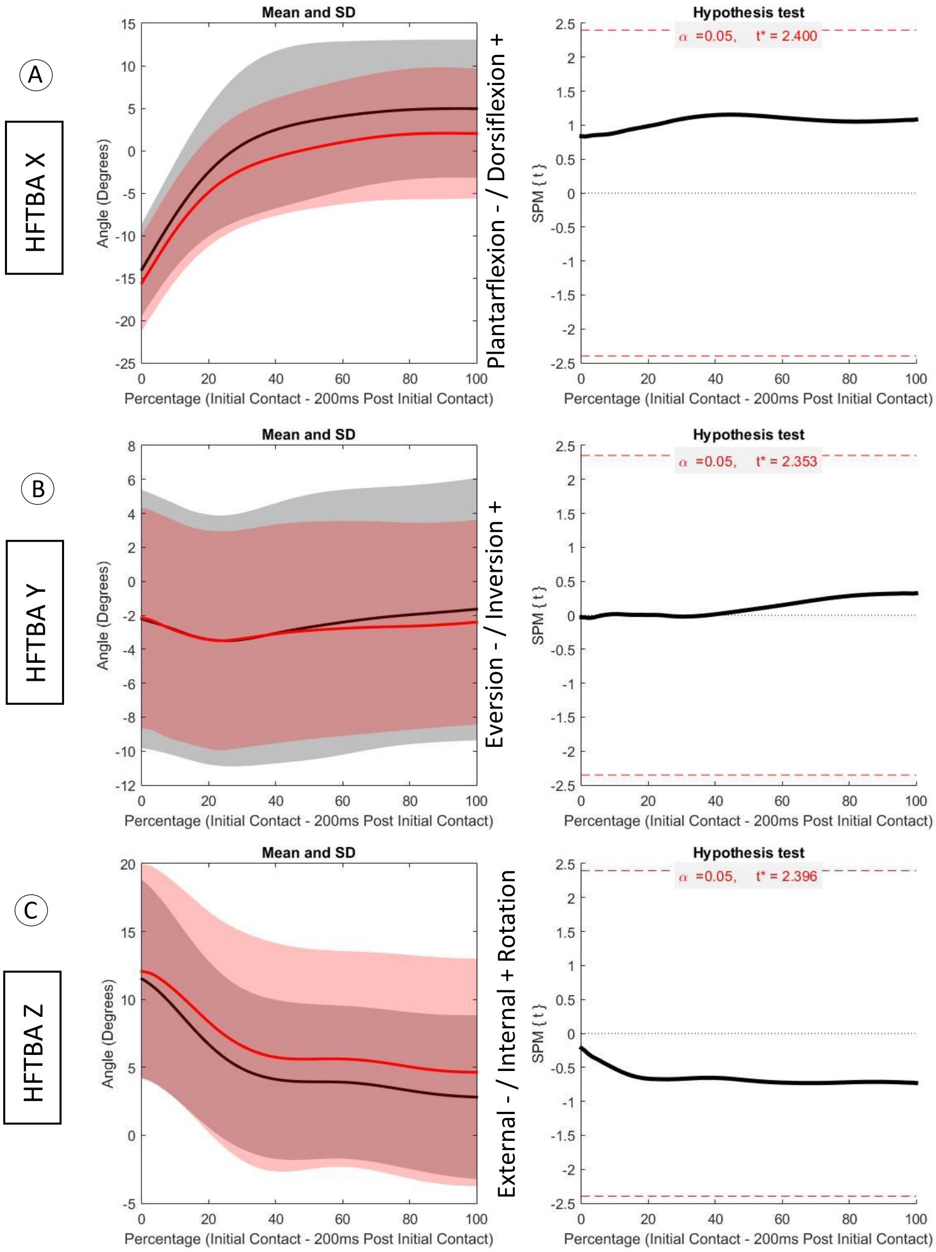


Figure H.38 Single leg landing - Hindfoot-Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

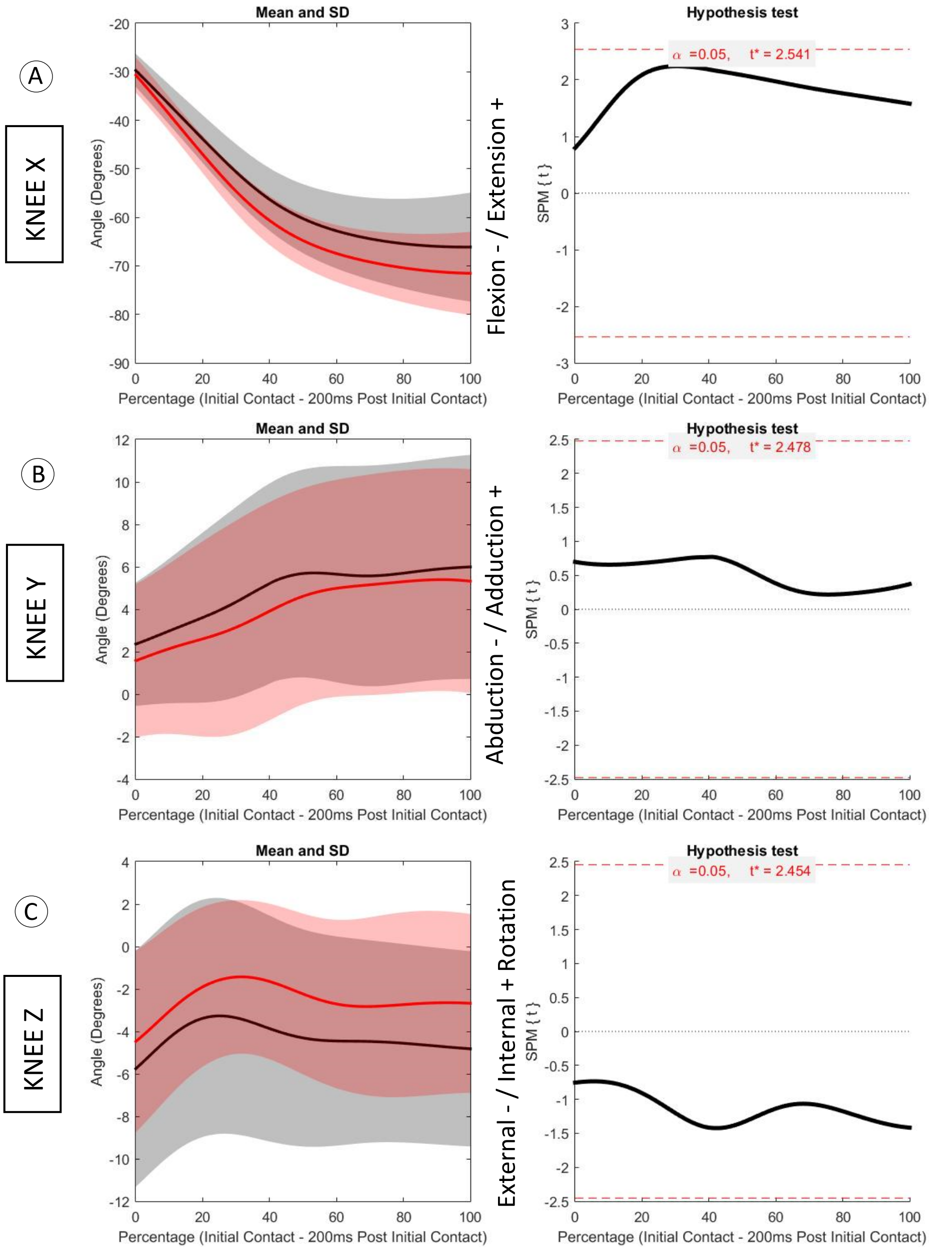


Figure H.39 Single leg landing - Knee angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

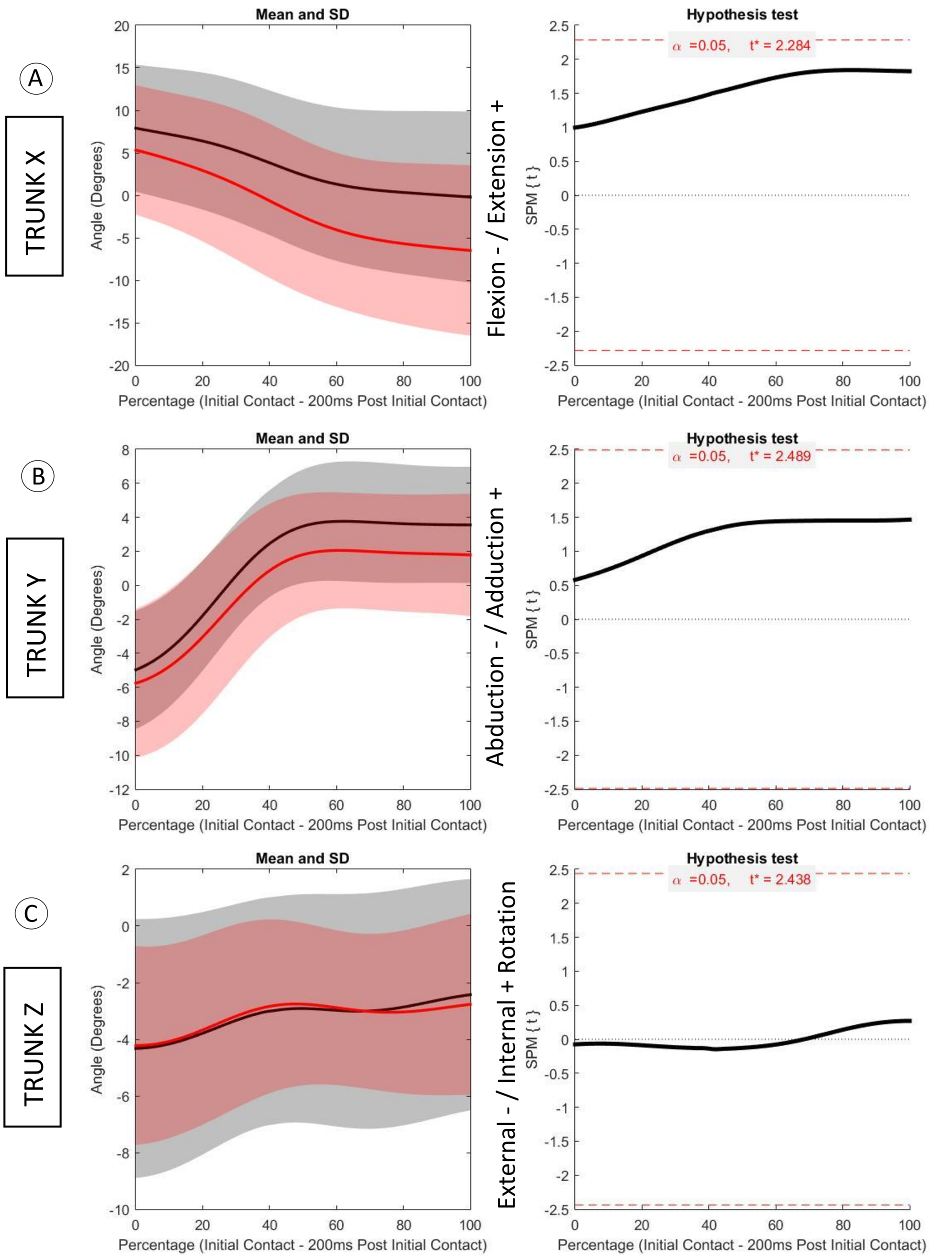


Figure H.40 Single leg landing - Trunk angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

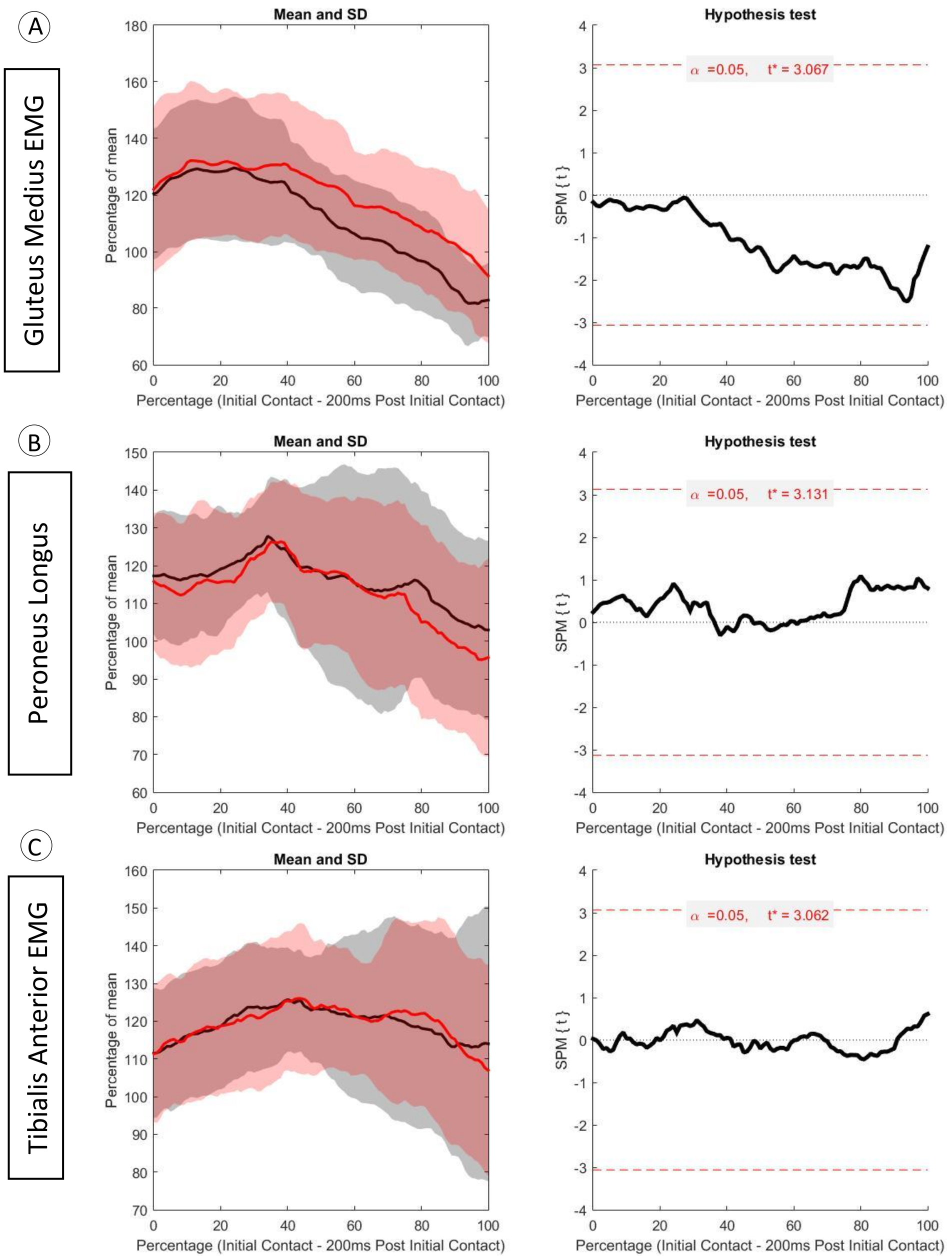


Figure H.41 Single leg landing - Gluteus Medius, Peroneus Longus and Tibialis Anterior muscle activation – IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

Appendix I Study 2 Part 2 – Landing Discrete

Angular Displacement

Table I.1 Peak angular displacement from 200 ms pre-initial contact to initial contact (pre-landing phase) (degrees)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	9.86	9.61	9.01	9.13	10.69	9.51	10.70	6.91
	Plantarflexion	-5.55	8.98	-6.76	10.72	-4.65	10.47	-4.79	9.02
FFHFA Frontal	Inversion	8.24	10.01	5.60	9.74	7.67	10.50	3.69	7.43
	Eversion	2.97	8.58	1.04	8.42	1.74	9.55	-1.54	7.38
FFHFA Transverse	Adduction	0.08	5.55	-0.70	7.39	-2.33	5.85	0.70	6.84
	Abduction	-6.30	6.73	-6.21	7.70	-8.38	6.50	-4.92	7.00
FFTBA Sagittal	Dorsiflexion	-1.42	7.06	-0.46	5.97	-0.63	8.35	-1.15	6.01
	Plantarflexion	-31.11	5.26	-31.19	7.28	-30.60	8.99	-30.30	7.20
FFTBA Frontal	Inversion	3.80	5.37	6.17	6.41	6.75	5.77	3.60	4.03
	Eversion	-1.70	5.05	-0.03	5.03	-0.29	4.86	-2.59	4.57
FFTBA Transverse	Internal Rotation	13.93*	5.53	10.89	4.75	10.04*	5.72	11.56	6.41
	External Rotation	4.88	5.90	1.42	5.56	2.44	5.99	2.95	7.11
HFTBA Sagittal	Dorsiflexion	-9.10	5.64	-8.08	6.34	-9.48	7.28	-10.84	7.52
	Plantarflexion	-23.76	5.88	-23.58	6.31	-24.44	5.88	-25.04	6.23
HFTBA Frontal	Inversion	-3.55	9.25	1.28	7.87	-0.46	9.17	0.90	6.55
	Eversion	-7.00	9.34	-2.74	8.00	-4.88	9.79	-2.88	6.38
HFTBA Transverse	Internal Rotation	15.63	6.26	12.34	7.62	13.54	7.93	12.55	7.98
	External Rotation	9.62	6.13	6.04	7.11	8.65	7.12	6.43	8.24
HIP Sagittal	Extension	38.06	7.48	38.30	7.51	40.18	8.62	39.48	8.77
	Flexion	30.89	7.30	31.11	7.29	33.08	6.50	32.20	6.95
HIP Frontal	Adduction	-7.74	4.00	-6.30	5.39	-8.01	3.91	-8.71	4.97
	Abduction	-15.58	4.16	-13.90	5.79	-14.61	4.01	-15.08	4.27
HIP Transverse	Internal Rotation	-2.66	5.90	-2.66	8.38	-2.21	6.72	-3.38	8.65
	External Rotation	-11.24	6.87	-11.72	8.77	-10.90	6.99	-11.68	9.52
KNEE Sagittal	Extension	-17.06	5.19	-17.12	4.38	-17.69	3.87	-17.71	4.47
	Flexion	-50.82	12.81	-52.85	10.29	-56.16	11.25	-56.10	9.71
KNEE Frontal	Adduction	3.12	2.08	4.07	3.62	3.74	4.46	2.99	3.98
	Abduction	0.27	2.05	0.21	2.95	-0.38	4.80	-0.87	4.38
KNEE Transverse	Internal Rotation	-4.09	4.79	-4.57	5.21	-3.28	4.22	-3.22	4.30
	External Rotation	-9.34	4.47	-9.40	5.82	-8.10	4.34	-7.75	4.65
TRUNK Sagittal	Extension	10.56	6.09	10.31	6.98	8.61	6.22	7.73	7.01
	Flexion	7.56	6.60	7.48	7.49	4.49	6.71	4.15	7.73
TRUNK Frontal	Adduction	-3.24	1.93	-1.84	2.89	-1.52	2.79	-2.48	4.07
	Abduction	-7.45	2.46	-5.68	3.54	-5.90	2.56	-6.61	4.32
TRUNK Transverse	Internal Rotation	0.16	4.18	-1.36	4.03	-0.28	4.11	-0.17	3.20
	External Rotation	-3.56	3.93	-4.64	4.66	-5.98	4.22	-4.54	3.53

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table I.2 Angular displacement at initial contact (degrees)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	7.94	8.10	6.37	9.55	8.08	9.38	8.44	7.99
FFHFA Frontal	4.02	8.53	2.15	8.35	3.57	9.78	0.02	7.13
FFHFA Transverse	-5.29	5.85	-5.06	7.57	-7.31	6.52	-3.74	6.87
FFTBA Sagittal	-8.22	3.14	-8.63	6.14	-8.97	6.79	-7.74	6.05
FFTBA Frontal	0.04	4.11	1.38	4.75	1.46	4.80	-0.36	3.98
FFTBA Transverse	10.62	5.71	7.16	4.89	6.98	5.27	8.71	5.85
HFTBA Sagittal	-14.16	5.59	-13.99	5.61	-15.34	5.92	-15.57	5.78
HFTBA Frontal	-6.16	9.33	-2.22	7.82	-4.17	9.75	-2.14	6.69
HFTBA Transverse	15.08	6.42	11.51	7.51	13.14	7.82	12.06	8.15
HIP Sagittal	33.99	7.78	33.54	7.94	35.66	6.73	34.74	7.16
HIP Frontal	-14.61	4.67	-11.88	4.96	-13.85	4.02	-14.59	4.22
HIP Transverse	-8.26	6.45	-8.46	8.85	-7.32	6.07	-8.67	8.72
KNEE Sagittal	-29.78	3.42	-29.72	3.51	-29.74	3.55	-30.68	3.74
KNEE Frontal	2.09	2.02	2.36	2.99	1.98	4.15	1.57	3.71
KNEE Transverse	-5.00	5.35	-5.75	5.73	-4.60	4.88	-4.47	4.41
TRUNK Sagittal	8.13	6.83	7.89	7.68	5.88	6.72	5.32	7.84
TRUNK Frontal	-6.84	2.40	-4.98	3.58	-5.40	2.53	-5.77	4.51
TRUNK Transverse	-3.29	3.94	-4.32	4.70	-5.55	4.42	-4.22	3.61

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table I.3 Peak angular displacement from initial contact to 200 ms post initial contact (impact phase) (degrees)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	23.84	8.04	21.47	9.94	24.31	9.30	25.53	7.90
	Plantarflexion	7.91	8.10	5.97	10.32	8.08	9.38	8.41	8.01
FFHFA Frontal	Inversion	4.56	8.56	2.72	8.02	4.00	9.78	1.02	7.03
	Eversion	-2.34	7.45	-3.35	6.85	-2.91	8.18	-5.77	7.17
FFHFA Transverse	Adduction	-5.16	5.88	-4.96	7.59	-6.94	6.35	-3.37	6.85
	Abduction	-12.49	7.33	-12.00	8.62	-14.35	7.41	-10.57	6.75
FFTBA Sagittal	Dorsiflexion	3.89	7.97	6.10	9.62	3.83	7.63	2.89	7.87
	Plantarflexion	-14.21	5.62	-14.01	5.63	-15.35	5.94	-15.58	5.78
FFTBA Frontal	Inversion	-3.65	9.13	-0.17	7.97	-2.29	9.60	-0.25	6.38
	Eversion	-7.84	9.71	-4.66	7.90	-6.69	9.98	-5.47	6.28
FFTBA Transverse	Internal Rotation	15.18	6.34	11.65	7.37	13.32	7.80	12.48	7.78
	External Rotation	5.81	5.86	1.61	6.02	4.63	6.79	3.26	8.46
HFTBA Sagittal	Dorsiflexion	4.33	7.69	6.10	9.62	3.79	7.59	2.89	7.87
	Plantarflexion	-13.05	6.54	-14.01	5.63	-14.45	7.88	-15.58	5.78
HFTBA Frontal	Inversion	-2.72	8.85	-0.17	7.97	-1.68	10.35	-0.25	6.38
	Eversion	-6.78	9.56	-4.66	7.90	-5.99	10.88	-5.47	6.28
HFTBA Transverse	Internal Rotation	14.83	6.59	11.65	7.37	12.94	8.11	12.48	7.78
	External Rotation	5.37	5.87	1.61	6.02	4.64	6.79	3.26	8.46
HIP Sagittal	Extension	52.23	11.95	53.91	15.92	58.71	13.80	58.15	14.04
	Flexion	33.75	7.93	33.20	8.53	35.66	6.73	34.69	7.15
HIP Frontal	Adduction	2.73	6.08	4.31 †	4.81	0.35	5.22	-0.17 †	4.07
	Abduction	-14.91	4.33	-12.72	5.86	-14.06	3.91	-14.67	4.28
HIP Transverse	Internal Rotation	1.95	5.70	2.39	8.67	3.24	7.96	2.00	8.12
	External Rotation	-8.17	6.44	-9.38	9.09	-7.63	6.13	-8.93	8.69
KNEE Sagittal	Extension	-29.78	3.42	-29.72	3.51	-29.74	3.55	-30.68	3.74
	Flexion	-66.97	8.44	-67.17	10.48	-72.37	7.43	-72.11	8.14
KNEE Frontal	Adduction	5.53	3.57	7.28	5.02	7.21	5.62	6.36	4.79
	Abduction	1.17	2.14	1.58	3.36	1.10	5.62	0.92	4.49
KNEE Transverse	Internal Rotation	-2.18	4.70	-2.58	5.40	-0.92	3.89	-0.57	3.99
	External Rotation	-6.30	4.88	-6.88	5.26	-5.74	4.74	-5.43	4.27
TRUNK Sagittal	Extension	8.20	6.87	8.17	7.74	5.92	6.74	5.38	7.88
	Flexion	-1.14	9.11	-0.65	10.04	-7.19	8.83	-6.65	10.24
TRUNK Frontal	Adduction	3.06	2.71	4.66	3.26	4.17	2.89	2.97	3.56
	Abduction	-6.84	2.40	-5.13	3.44	-5.42	2.56	-5.97	4.23
TRUNK Transverse	Internal Rotation	-0.55	3.21	-1.43	4.20	-2.15	4.10	-1.40	3.05
	External Rotation	-4.09	3.62	-5.30	4.29	-6.46	4.48	-5.41	3.24

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Angular Velocity

Table I.4 Peak angular velocity from 200 ms pre-initial contact to initial contact (pre-landing phase) (degrees/second)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	373.63	75.68	362.66	89.80	366.50	93.12	382.62	81.80
	Plantarflexion	-136.11	50.94	-158.66	68.17	-146.34	50.82	-167.63	60.09
FFHFA Frontal	Inversion	57.46	21.75	53.03	28.01	64.56	23.92	65.06	18.13
	Eversion	-107.78	59.87	-86.46	51.29	-97.47	48.96	-96.27	46.56
FFHFA Transverse	Adduction	75.79	31.95	58.65	30.95	68.20	25.24	66.40	33.27
	Abduction	-164.56	81.44	-152.67	73.36	-169.56	83.11	-153.38	60.91
FFTBA Sagittal	Dorsiflexion	707.23	97.76	696.56	77.48	681.98	88.39	701.68	58.73
	Plantarflexion	-280.46	95.90	-312.59	71.65	-294.81	108.28	-312.13	73.49
FFTBA Frontal	Inversion	70.23	29.41	75.65	56.00	88.42	30.37	83.25	30.57
	Eversion	-80.99	34.62	-98.10	59.95	-117.00	69.48	-101.75	54.38
FFTBA Transverse	Internal Rotation	114.46	34.61	105.87	39.32	97.26	41.35	104.95	42.48
	External Rotation	-167.91	50.45	-187.83	60.90	-143.15	54.78	-153.68	44.03
HFTBA Sagittal	Dorsiflexion	303.68	55.15	326.78	91.55	293.52	49.42	323.09	84.93
	Plantarflexion	-149.42	47.95	-166.75	30.84	-158.12	54.89	-155.07	41.90
HFTBA Frontal	Inversion	41.79	19.02	44.99	24.93	47.83	23.48	49.26	21.16
	Eversion	-49.84*	24.86	-63.25	25.66	-70.73*	35.41	-62.99	41.44
HFTBA Transverse	Internal Rotation	78.86	22.86	81.73	23.56	82.69	32.64	89.20	34.42
	External Rotation	-56.39	24.21	-70.36 †	34.76	-46.81	30.52	-49.98 †	29.93
HIP Sagittal	Extension	93.70	46.03	93.14	37.99	90.40	37.22	89.37	57.42
	Flexion	-54.96	44.42	-59.74	39.45	-62.68	40.76	-67.53	36.15
HIP Frontal	Adduction	33.88	22.52	32.08	22.60	38.84	24.54	39.77	24.82
	Abduction	-58.47	25.34	-43.97	18.71	-66.55	36.59	-64.03	37.88
HIP Transverse	Internal Rotation	123.48	48.17	134.41	38.62	121.90	45.33	108.83	43.34
	External Rotation	-111.09	44.63	-111.93	37.31	-101.51	22.66	-103.39	31.72
KNEE Sagittal	Extension	371.11	95.32	371.05	75.30	386.93	76.63	396.34	80.61
	Flexion	-335.49	70.59	-331.71	69.12	-371.38	45.43	-370.67	62.77
KNEE Frontal	Adduction	57.16*	21.55	70.21	29.95	75.41*	28.51	70.47	30.82
	Abduction	-29.14	23.91	-39.64	33.91	-42.58	32.52	-41.32	28.90
KNEE Transverse	Internal Rotation	72.22	30.95	80.24	34.08	71.24	34.43	70.85	25.72
	External Rotation	-48.63	30.55	-48.73	34.29	-50.92	34.03	-45.17	30.09
TRUNK Sagittal	Extension	17.82	16.86	19.96 †	13.04	30.68	24.15	32.49 †	14.67
	Flexion	-40.63*	23.41	-43.23	23.94	-62.01*	30.80	-54.86	29.89
TRUNK Frontal	Adduction	37.62	21.00	40.23	17.29	32.06	19.73	32.00	17.34
	Abduction	-38.50	16.04	-35.19	11.74	-40.60	23.07	-43.43	21.81
TRUNK Transverse	Internal Rotation	20.33	20.22	13.83	17.21	13.69	18.29	24.70	19.51
	External Rotation	-41.30	29.78	-34.15	13.09	-51.69	16.07	-36.82	25.50

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table I.5 Angular velocity at initial contact (degrees/second)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	369.31	77.23	352.99	92.84	362.01	93.83	366.14	101.52
FFHFA Frontal	-90.85	79.12	-68.48	68.60	-79.51	64.63	-61.97	73.65
FFHFA Transverse	-177.30	84.65	-150.15	76.15	-170.52	80.76	-147.77	63.06
FFTBA Sagittal	707.23	97.76	696.56	77.48	681.98	88.39	701.68	58.73
FFTBA Frontal	-34.47	78.22	-73.95	78.57	-84.09	98.56	-59.54	90.97
FFTBA Transverse	-176.85	37.34	-187.83	60.90	-149.70	49.90	-153.52	44.20
HFTBA Sagittal	306.24	52.32	340.17	87.04	291.14	53.00	322.49	85.16
HFTBA Frontal	16.24*	58.27	4.16	52.13	-44.19*	57.79	-35.37	65.72
HFTBA Transverse	-5.14	57.24	-2.39	62.64	-13.14	48.56	-16.29	41.38
HIP Sagittal	45.99	35.25	45.31	53.80	55.33	41.60	53.37	60.52
HIP Frontal	-16.84	53.11	-4.37	34.15	-30.92	50.14	-24.85	49.84
HIP Transverse	118.28	58.98	134.39 †	38.65	118.03	47.66	103.90 †	49.57
KNEE Sagittal	-335.82	71.28	-330.16	69.82	-370.26	45.21	-369.94	62.97
KNEE Frontal	54.53	21.52	59.80	40.33	57.72	44.84	57.62	42.08
KNEE Transverse	34.17	44.12	44.94	48.37	23.02	60.07	38.22	46.17
TRUNK Sagittal	-31.20*	28.40	-34.78	30.00	-57.30*‡	33.50	-47.51 ‡	35.57
TRUNK Frontal	33.84	24.28	38.40	19.11	28.85	24.05	29.06	26.48
TRUNK Transverse	13.37	24.73	7.37	22.27	2.58	25.72	1.00	22.00

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table I.6 Peak angular velocity from initial contact to 200 ms post-initial contact (impact phase) (degrees/second)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	369.56	77.22	356.38	94.42	362.29	94.28	380.61	81.93
	Plantarflexion	-38.47	26.28	-45.57	51.44	-34.27	28.17	-39.29	34.78
FFHFA Frontal	Inversion	52.74	31.07	42.10	20.45	45.92	29.98	56.61	39.67
	Eversion	-117.06	59.41	-94.17	54.81	-101.22	57.64	-111.97	54.55
FFHFA Transverse	Adduction	48.12	26.10	36.84	17.60	39.50	19.34	43.14	21.01
	Abduction	-175.31	79.02	-158.75	71.23	-172.17	83.96	-166.17	55.05
FFTBA Sagittal	Dorsiflexion	707.23	97.76	696.56	77.48	681.98	88.39	701.68	58.73
	Plantarflexion	-16.70	30.92	-18.26	24.87	-17.56	28.96	-15.86	26.22
FFTBA Frontal	Inversion	58.87	22.49	47.88	19.20	46.74	27.56	39.66	21.61
	Eversion	-73.54	52.42	-104.03	62.13	-115.19	75.92	-94.30	71.60
FFTBA Transverse	Internal Rotation	42.20	29.76	48.89	26.71	48.42	38.33	47.74	39.46
	External Rotation	-216.96	47.64	-250.05	69.07	-191.78	65.27	-214.63	52.33
HFTBA Sagittal	Dorsiflexion	308.73	51.65	343.02	86.81	306.87	41.43	332.22	86.06
	Plantarflexion	-8.96	21.29	-17.26	37.50	-20.37	29.61	-21.23	21.96
HFTBA Frontal	Inversion	58.39	29.44	49.27	27.38	44.15	27.75	64.35	45.75
	Eversion	-59.88	35.66	-66.10	38.87	-72.10	41.04	-82.59	70.96
HFTBA Transverse	Internal Rotation	38.32	35.23	41.47	19.22	49.88	48.04	49.55	41.30
	External Rotation	-122.19	37.85	-145.56	41.52	-104.36	55.74	-131.27	70.13
HIP Sagittal	Extension	157.41	38.73	175.57	71.06	183.00	69.33	182.14	67.13
	Flexion	3.29	37.55	6.45	49.85	23.45	39.81	18.14	40.28
HIP Frontal	Adduction	110.16	57.07	103.38	51.49	92.70	39.80	89.90	44.57
	Abduction	-39.30	34.65	-46.14	55.92	-54.77	46.24	-52.11	41.44
HIP Transverse	Internal Rotation	232.03	53.61	246.46	73.24	215.53	72.09	218.62	60.96
	External Rotation	-25.32	29.29	-43.62	38.65	-43.18	42.53	-25.29	36.25
KNEE Sagittal	Extension	7.25	57.30	3.96	61.94	-13.85	47.55	-5.42	46.76
	Flexion	-374.59*	56.94	-368.93 †	58.19	-425.34*	48.30	-418.13 †	74.62
KNEE Frontal	Adduction	62.19	23.99	74.51	38.59	84.03	48.56	77.45	38.06
	Abduction	-33.73	20.17	-30.17	20.12	-34.48	37.09	-39.64	27.42
KNEE Transverse	Internal Rotation	58.83	34.86	70.28	36.11	63.23	42.91	73.00	42.96
	External Rotation	-64.60*	28.76	-74.46	42.48	-67.37*	59.56	-67.69	54.86
TRUNK Sagittal	Extension	2.50	22.43	5.94	24.55	-6.22	22.00	-6.55	18.21
	Flexion	-92.30	32.88	-84.63 †	29.50	-115.77	38.10	-114.77 †	26.70
TRUNK Frontal	Adduction	120.73	45.43	119.27 †	45.32	116.84 ‡	35.56	71.09 †‡	57.63
	Abduction	-21.35	16.94	-17.78 †	18.01	-15.13 ‡	16.54	-66.29 †‡	53.51
TRUNK Transverse	Internal Rotation	55.38	26.26	45.75	24.88	59.81 ‡	33.81	40.94 ‡	22.19
	External Rotation	-29.25	20.50	-19.68 †	9.65	-36.71	28.93	-33.68 †	19.11

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Angular Acceleration

Table I.7 Peak angular acceleration from 200 ms pre-initial contact to initial contact (pre-landing) (degrees/second²)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	+ VE	9454.97	1941.66	9443.01	1988.07	9609.80	2395.61	9516.02	1976.01
	- VE	-2422.74	710.53	-3008.69	1341.57	-2591.74	1087.98	-4263.24	4268.91
FFHFA Frontal	+ VE	1753.13	832.82	1667.26	791.52	1892.34	846.40	2158.57	1276.54
	- VE	-2571.56	940.44	-2720.08	2450.26	-2952.60	1872.88	-3440.03	3794.72
FFHFA Transverse	+ VE	1806.72	830.80	1769.26	753.32	1604.92	745.87	1975.51	1349.74
	- VE	-3875.86	1847.77	-3446.95	1540.79	-3839.43	1818.68	-4267.89	3642.15
FFTBA Sagittal	+ VE	16741.81	2083.54	16572.84	1792.58	16549.20	2467.55	16624.39	1796.91
	- VE	-4137.28	1609.16	-4768.80	1541.21	-4510.47	1954.39	-5038.67	1885.24
FFTBA Frontal	+ VE	2286.84	1191.17	2110.14	1597.95	2140.41	998.44	2483.30	1116.77
	- VE	-2416.90	799.81	-2513.54	1206.44	-2794.68	1293.84	-2852.62	1210.97
FFTBA Transverse	+ VE	3108.99	1505.64	3236.50	1609.62	2760.64	1714.30	3292.39	1903.20
	- VE	-4997.34	1465.07	-5800.12	1957.87	-4532.27	1632.92	-5230.08	1267.57
HFTBA Sagittal	+ VE	7140.97	1166.04	8015.07	2086.05	6967.61	1223.23	7437.20	1417.84
	- VE	-2392.53	1005.94	-3068.64	1509.40	-2562.90	1019.92	-3574.89	4694.13
HFTBA Frontal	+ VE	1505.50	547.15	1884.27	1563.77	1537.16	1264.98	1560.75	621.23
	- VE	-1628.81	910.40	-1700.29	696.49	-1743.30	905.17	-2390.15	1934.86
HFTBA Transverse	+ VE	2514.73	1058.00	2593.69	995.10	2592.65	1209.12	2654.16	1389.41
	- VE	-3735.46	1248.56	-4621.53	1796.26	-3905.84	1929.40	-4869.23	3297.69
HIP Sagittal	+ VE	2353.51	899.51	2309.04	890.42	2833.92	1210.26	2828.31	1260.25
	- VE	-1879.33	1157.83	-1893.90	1094.99	-1770.13	1382.41	-1929.62	1220.16
HIP Frontal	+ VE	1762.64	848.28	1557.30	710.47	1554.56	625.69	1704.60	742.87
	- VE	-1286.87 *	509.74	-1251.97	828.40	-1904.37 *	1079.25	-1656.68	788.02
HIP Transverse	+ VE	4486.11	1292.25	5127.08	1672.09	4110.83	1211.49	4216.73	1279.66
	- VE	-1605.40	776.91	-1490.93 †	758.46	-1609.80	624.45	-1696.98 †	748.97
KNEE Sagittal	+ VE	5265.32	3687.36	5550.62	3104.69	5785.92	2939.91	5780.79	2768.68
	- VE	-6780.10	2936.57	-6750.67	2950.78	-6267.32	4265.71	-6206.08	4523.21
KNEE Frontal	+ VE	1372.68	584.13	1604.87	692.45	1627.40	703.79	1555.26	637.11
	- VE	-1147.12	814.92	-1213.41	660.56	-1174.28	645.68	-1304.41	707.14
KNEE Transverse	+ VE	2365.95	1090.00	2616.47	1091.95	2466.21	1168.20	2676.11	1220.34
	- VE	-1733.61	851.15	-1946.25	1210.50	-1597.37	1554.50	-1671.59	1178.58
TRUNK Sagittal	+ VE	680.76	389.55	727.76	330.69	799.77 ‡	316.71	980.38 ‡	472.68
	- VE	-992.85 *	603.86	-917.44 †	463.71	-1430.11 *	613.83	-1496.08 †	814.68
TRUNK Frontal	+ VE	1996.31	840.24	1865.17	573.97	1801.35	607.48	1771.54	679.27
	- VE	-717.97	431.78	-592.62	204.58	-635.98	295.50	-781.26	529.04
TRUNK Transverse	+ VE	1471.28	1373.24	930.88	408.84	1143.49	366.25	1273.81	1002.66
	- VE	-929.65	2135.99	-448.44	184.80	-644.41	309.34	-1045.49	1709.85

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table I.8 Angular acceleration at initial contact (degrees/second^2)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	-1004.79	780.39	-1684.31	1688.27	-781.76	1055.07	-1129.38	1195.92
FFHFA Frontal	-82.25	1212.87	246.13	1108.00	13.10	1507.59	376.44	1980.83
FFHFA Transverse	42.94	810.47	-209.23	825.18	323.85	857.53	185.24	1230.85
FFTBA Sagittal	279.60	646.62	447.51	723.07	572.77	648.99	491.32	673.75
FFTBA Frontal	-663.97	1057.63	-363.13	1092.47	-67.72	1535.64	-688.85	1024.38
FFTBA Transverse	-3422.90	1282.74	-4295.74	1650.10	-3684.36	1622.10	-3835.47	1774.52
HFTBA Sagittal	1237.64	960.58	2095.17	1730.86	1312.37	1350.58	1667.83	1742.53
HFTBA Frontal	-347.16	954.36	-288.65	999.12	29.98	1090.96	-795.10	1890.26
HFTBA Transverse	-3453.29	1184.80	-4107.60	1686.41	-3993.57	2139.16	-3897.24	1722.01
HIP Sagittal	1938.58	1060.99	2097.82	1136.79	2461.51	1524.41	2314.75	1609.51
HIP Frontal	1395.16	1066.85	1042.73	1409.84	928.94	1413.50	1339.16	982.94
HIP Transverse	4108.99	1338.28	4525.08	1849.07	3497.15	1771.04	3778.95	1473.04
KNEE Sagittal	-2087.70*	1095.33	-1710.22	1571.70	-3450.92*	2048.42	-2994.81	2216.20
KNEE Frontal	200.82	446.39	294.20	573.91	434.07	876.54	249.50	564.50
KNEE Transverse	408.16	1399.93	671.22	1567.96	1165.25	1347.19	1095.47	1983.45
TRUNK Sagittal	-277.57	738.83	-70.90 †	662.24	-716.06	861.38	-637.44 †	904.71
TRUNK Frontal	1938.23	755.14	1812.82	629.31	1772.02	613.45	1676.37	748.30
TRUNK Transverse	1112.42	421.51	820.67	440.18	941.29	414.60	845.29	528.05

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table I.9 Peak angular acceleration from initial contact to 200 ms post-initial contact (impact phase) (degrees/second^2)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	+ VE	1296.51	1382.54	1669.65	1970.85	1265.14	993.12	2844.31	5576.47
	- VE	-6430.93	2567.21	-6777.29	3054.35	-5863.13	2604.41	-6687.44	1946.98
FFHFA Frontal	+ VE	3734.24	2336.75	2784.92	1318.28	2695.75	1544.78	3955.56	2911.91
	- VE	-2692.54	4390.86	-1827.73	1395.77	-1708.79	1265.01	-3982.26	5270.29
FFHFA Transverse	+ VE	5307.68	5160.51	3481.63	1067.09	3901.93	1463.59	5305.06	4195.64
	- VE	-2075.50	3393.57	-1298.70	591.09	-1185.65	572.97	-2792.42	4098.33
FFTBA Sagittal	+ VE	1074.69	894.97	1044.44	720.63	983.77	492.27	916.30	504.09
	- VE	-10337.12	3104.36	-10545.24	1569.43	-9629.50	2964.98	-10684.92	1285.86
FFTBA Frontal	+ VE	3084.80	1269.56	3677.95	1434.21	3604.97	1541.29	3242.01	1457.56
	- VE	-1425.16	536.52	-1308.79	515.35	-1291.25	946.57	-1338.30	587.94
FFTBA Transverse	+ VE	4944.01	985.54	5708.02	1621.11	4969.71	2582.49	5072.94	2009.59
	- VE	-3639.03	1263.35	-4404.21	1487.44	-3485.96	1576.39	-4046.10	1390.70
HFTBA Sagittal	+ VE	2159.94	1735.68	2480.82	1581.40	2019.75	983.68	2308.96	1686.18
	- VE	-4813.07	1389.21	-6093.89	2584.54	-4947.49	1364.58	-7408.14	6055.93
HFTBA Frontal	+ VE	3454.84	3943.90	3145.65	1543.61	2730.36	1046.67	4464.90	4912.11
	- VE	-2242.69	2179.10	-1525.75	713.88	-1645.92	1107.71	-2686.00	2919.41
HFTBA Transverse	+ VE	3883.07	3415.51	3789.73	1270.06	3623.28	2452.62	4782.86	4180.95
	- VE	-5016.78	5509.86	-4532.06	1327.29	-4075.92	1918.14	-5523.68	3919.93
HIP Sagittal	+ VE	3059.01	1158.91	3131.51	1112.52	3471.94	1724.85	3416.89	1288.13
	- VE	-2790.98	1224.48	-3528.56	3282.62	-2828.23	1484.64	-2673.70	1389.20
HIP Frontal	+ VE	3026.37	1421.36	3553.50	3201.01	2753.89	1093.09	2656.85	946.56
	- VE	-2490.57	1144.67	-2670.65	1520.37	-2496.06	1303.07	-2656.87	1156.43
HIP Transverse	+ VE	4527.92	1526.97	4902.81	2315.36	4247.99	2076.64	4173.90	1455.75
	- VE	-4457.93	1367.70	-5311.99	3077.13	-3728.92	1649.57	-3854.65	1754.59
KNEE Sagittal	+ VE	4786.72	626.61	4740.66	1591.80	5266.73	1244.46	5295.15	1225.67
	- VE	-2188.67*	1188.25	-1976.56	1166.52	-3446.65*	1579.53	-2960.02	1854.74
KNEE Frontal	+ VE	1124.40	620.08	1154.95	788.95	1522.54	1101.44	1498.16	1062.09
	- VE	-2476.52	963.03	-2683.21	1610.88	-2498.33	1739.49	-2300.88	1397.06
KNEE Transverse	+ VE	2590.71	945.56	3161.85	2299.45	2780.97	2176.26	2872.82	2269.66
	- VE	-2555.46	1038.91	-2973.10	1417.74	-2323.75	1956.70	-2692.94	1985.28
TRUNK Sagittal	+ VE	1627.14	531.77	1459.30 †	519.16	1721.49	609.90	1822.69 †	515.05
	- VE	-1487.86	657.35	-1604.35	1533.28	-1731.67	1238.40	-1731.40	631.05
TRUNK Frontal	+ VE	2323.98	886.83	2298.82	990.63	2335.86	981.85	2238.38	881.79
	- VE	-2485.02	959.72	-2390.63	820.55	-2246.13	961.58	-2444.10	813.70
TRUNK Transverse	+ VE	1571.81	1212.36	1314.78	1147.54	1683.41	1566.05	1342.94	591.57
	- VE	-1716.48	871.56	-1316.00	960.31	-1626.28	759.57	-1338.28	399.41

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Appendix J Study 2 Part 3 – Landing Regressions

SPSS Correlation Outputs

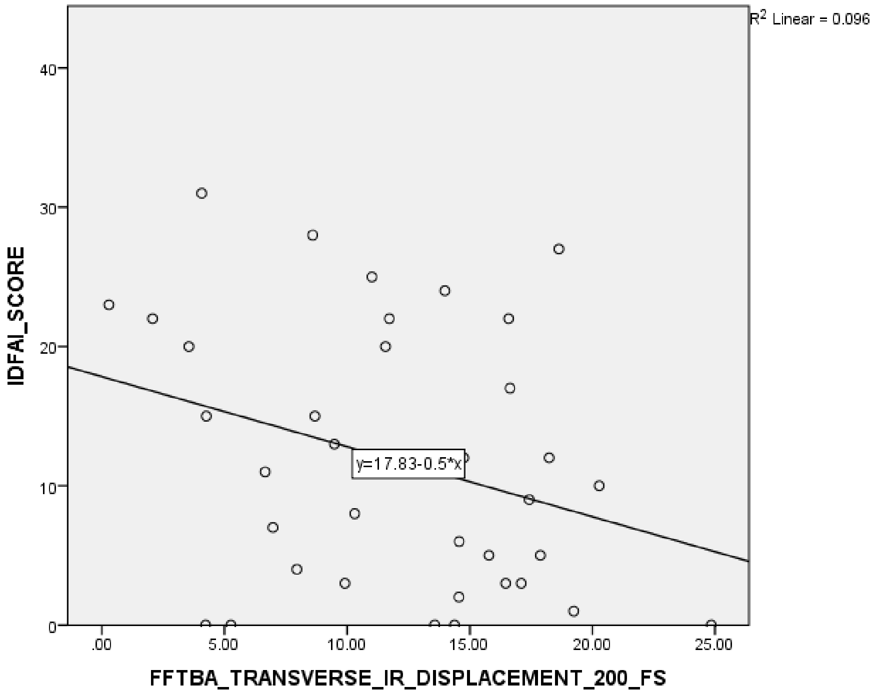


Figure J.1 Pearson's correlation FFTBA transverse internal rotation displacement 200 ms pre IC to IC and IdFAI score

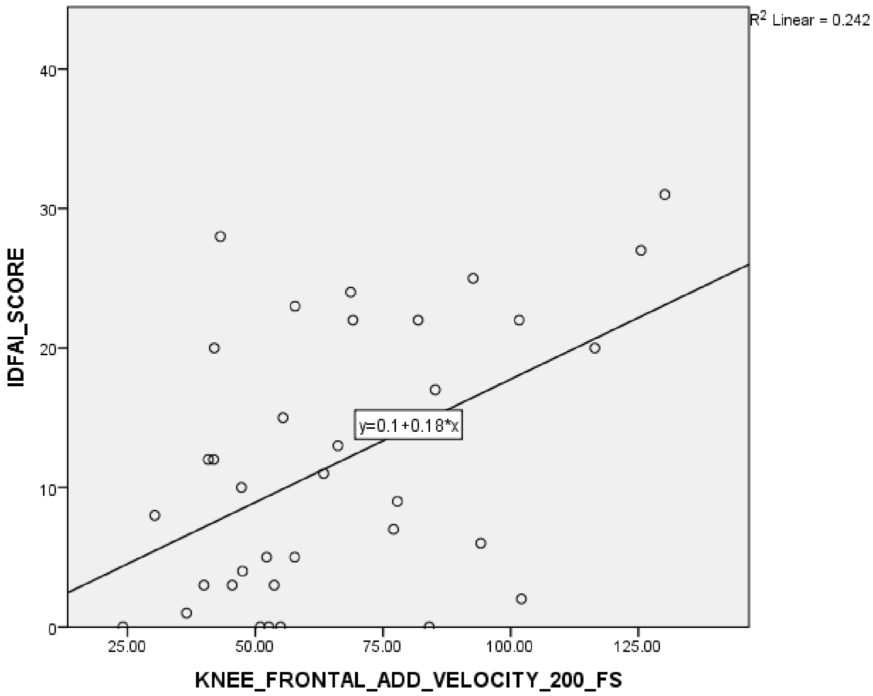


Figure J.2 Pearson's correlation knee frontal adduction velocity 200 ms pre IC to IC and IdFAI score

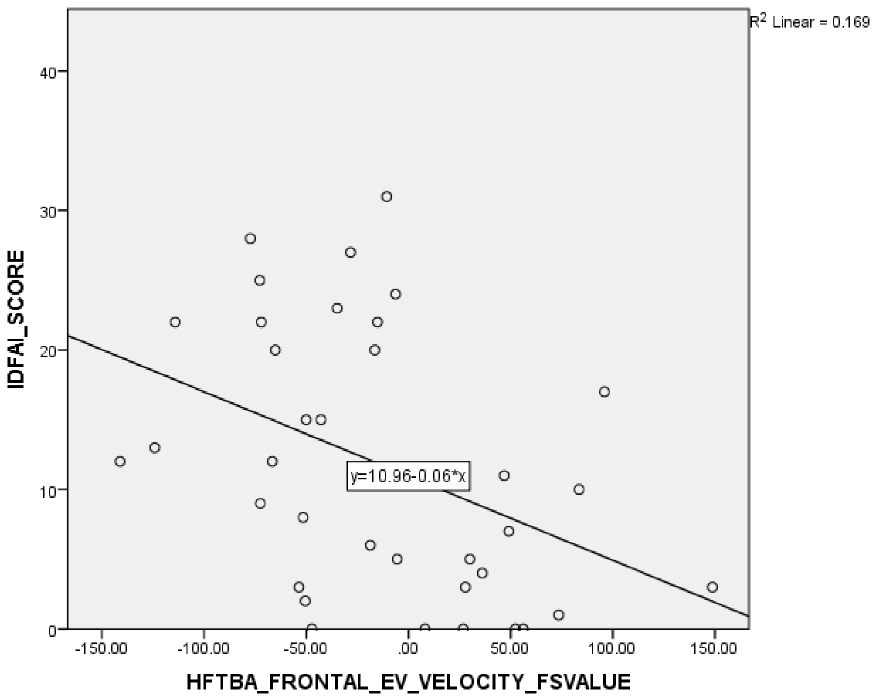


Figure J.3 Pearson's correlation HFTBA frontal eversion velocity at IC and IdFAI score

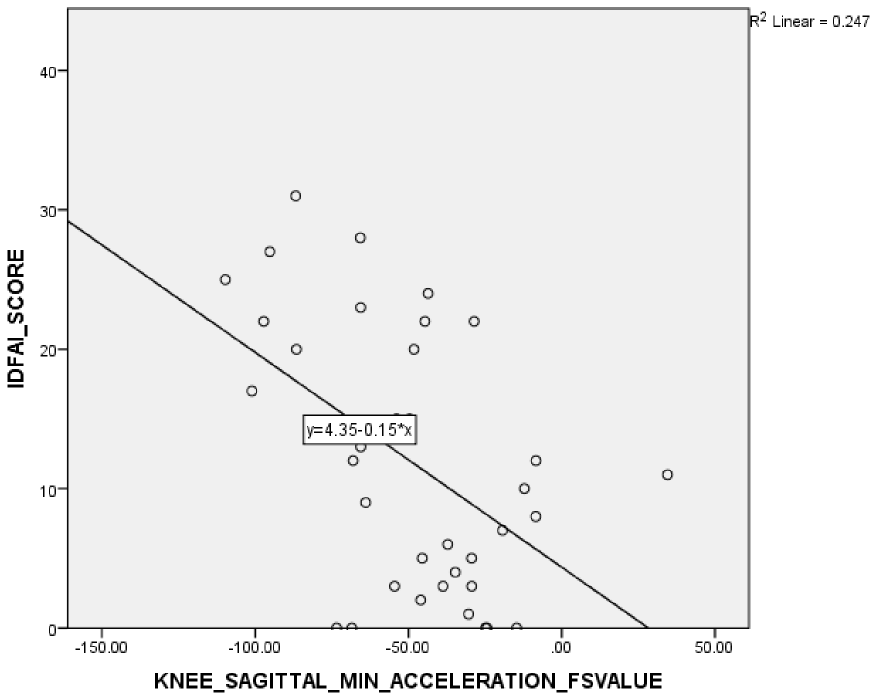


Figure J.4 Pearson's correlation knee sagittal flexion acceleration at IC and IdFAI score

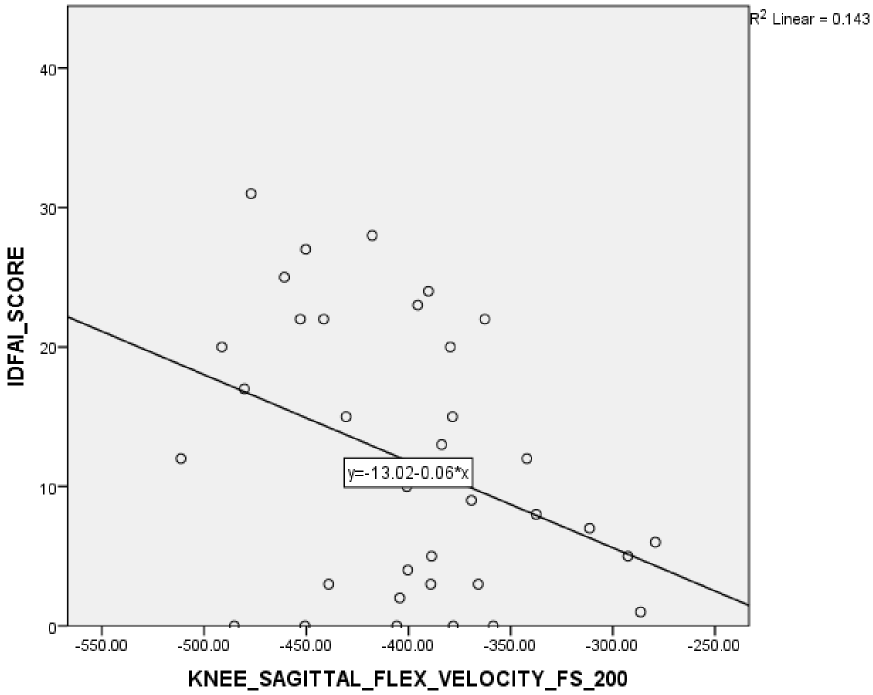


Figure J.5 Pearson's correlation knee sagittal flexion velocity IC to 200 ms post and IdFAI score

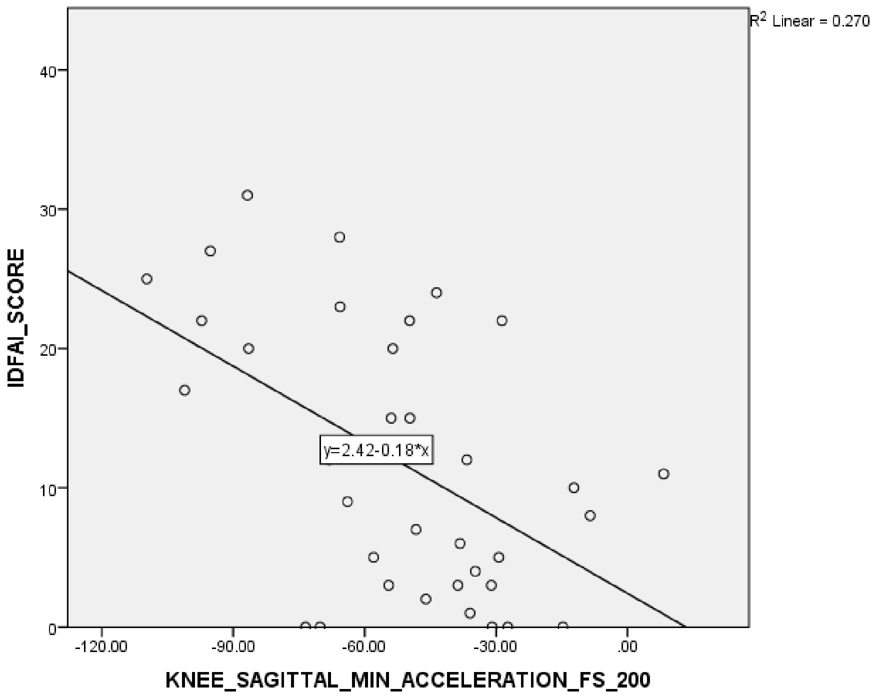


Figure J.6 Pearson's correlation knee sagittal peak negative acceleration IC to 200 ms post and IdFAI score

Table J.1 R values for regression analysis model with IdFAI as dependent variable

	R	R Square	Adj R Square	F	Sig
Knee sagittal min acceleration IC-200	0.520	0.270	0.249	12.576	0.001
Knee sagittal min acceleration IC-200, Knee frontal adduction velocity 200-IC	0.593	0.352	0.312	4.152	0.050

Appendix K Study 3 Part 1 – Cutting

Matched Control vs Affected limb of CAI group

- **200 ms pre initial contact to initial contact**

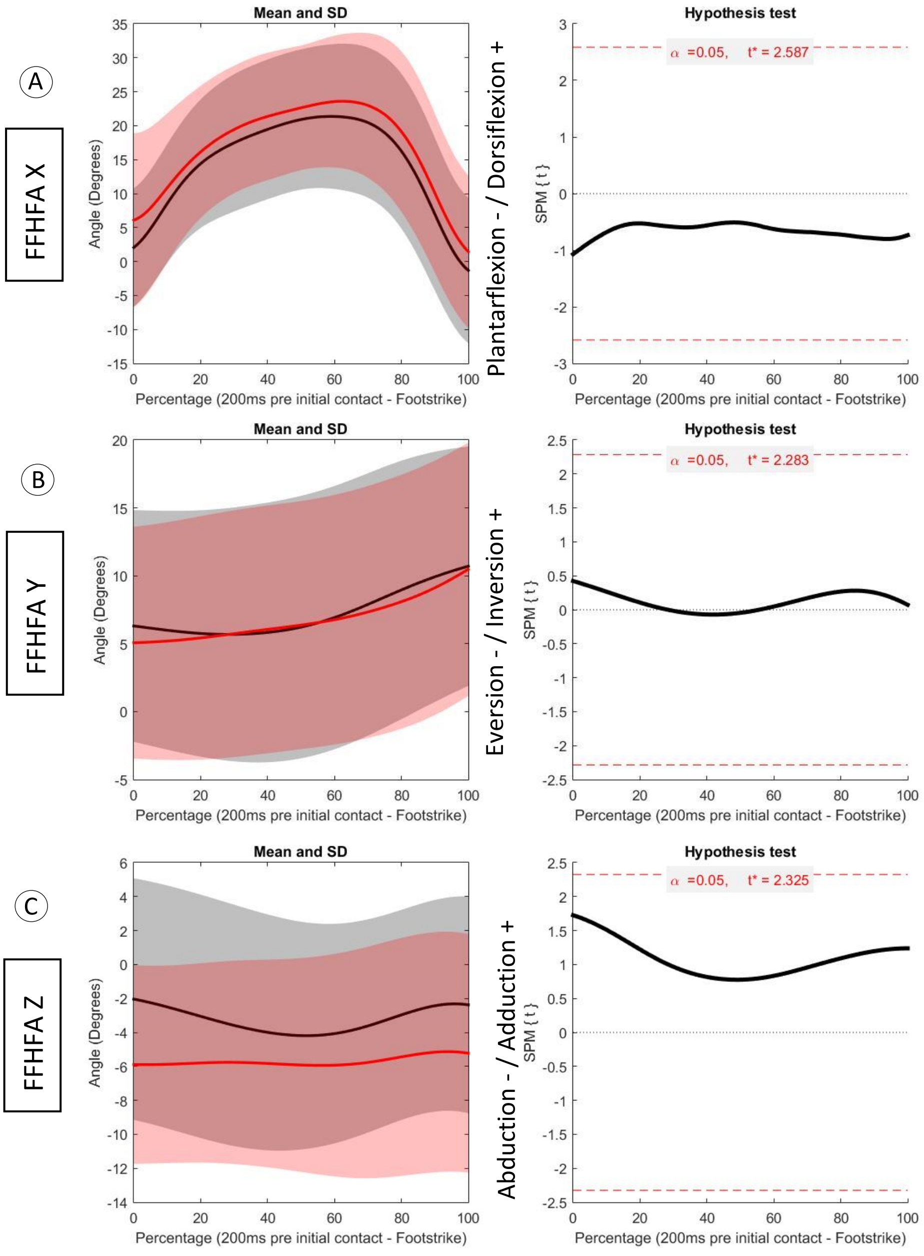


Figure K.1 Cutting- Forefoot – Hindfoot angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

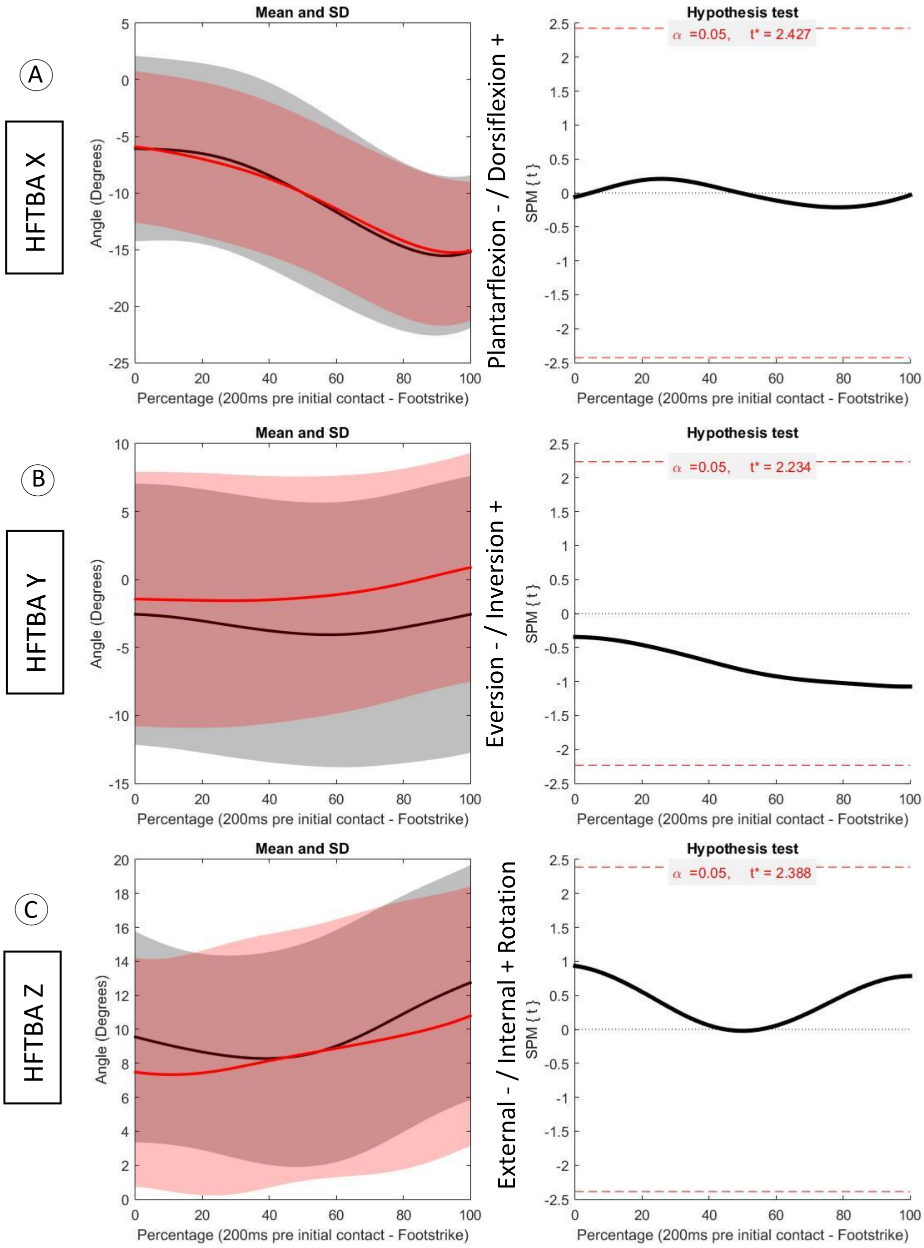


Figure K.2 Cutting- Hindfoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

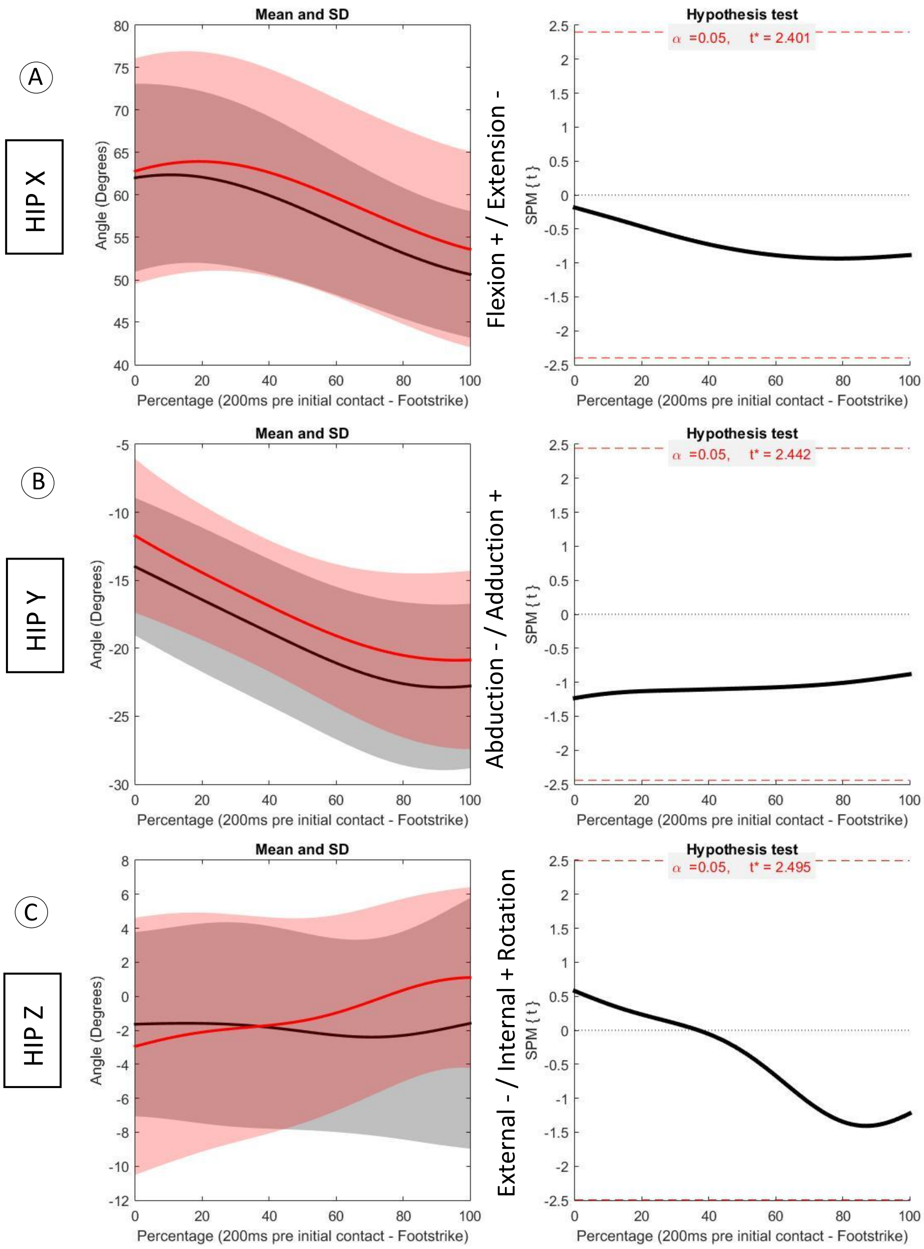


Figure K.3 Cutting- Hip angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

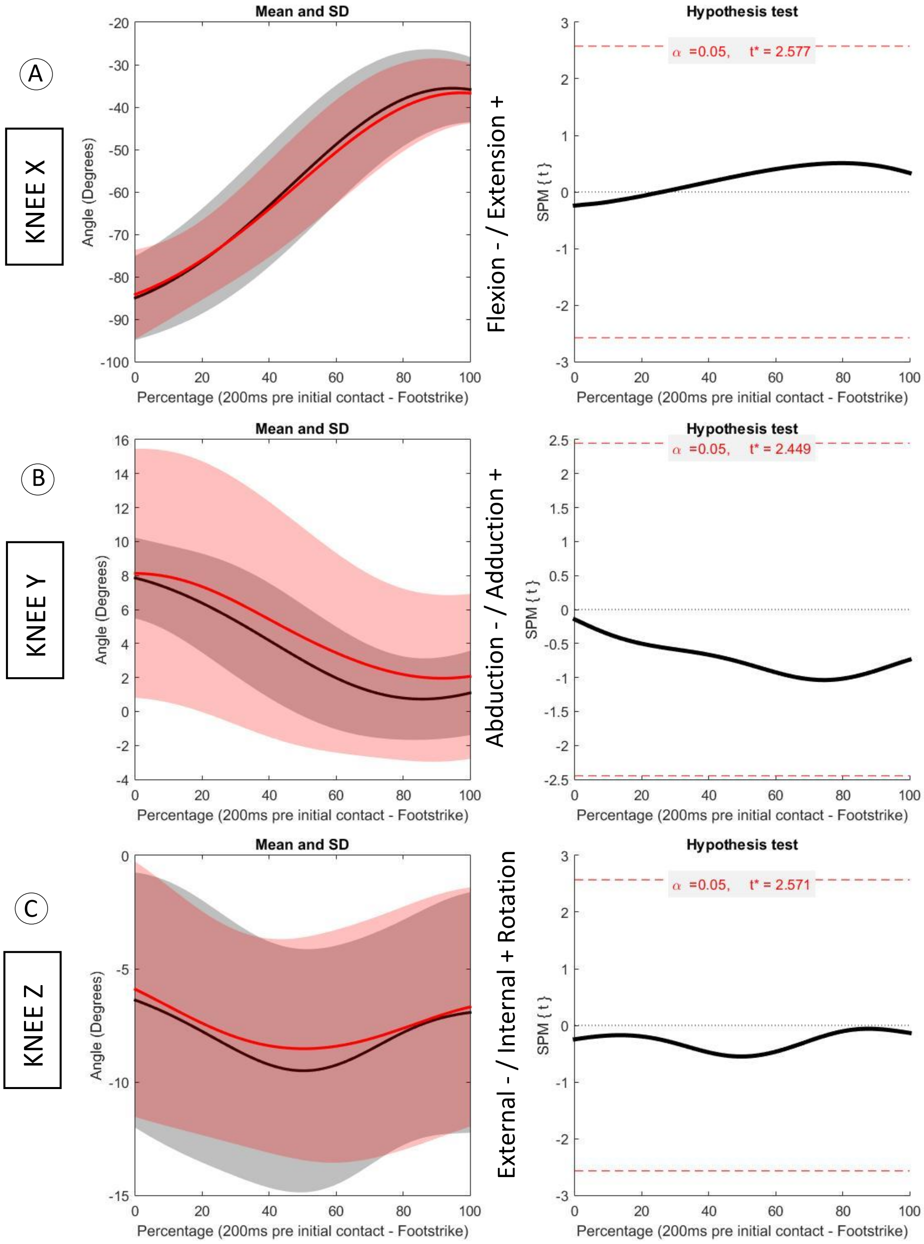


Figure K.4 Cutting- Knee angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

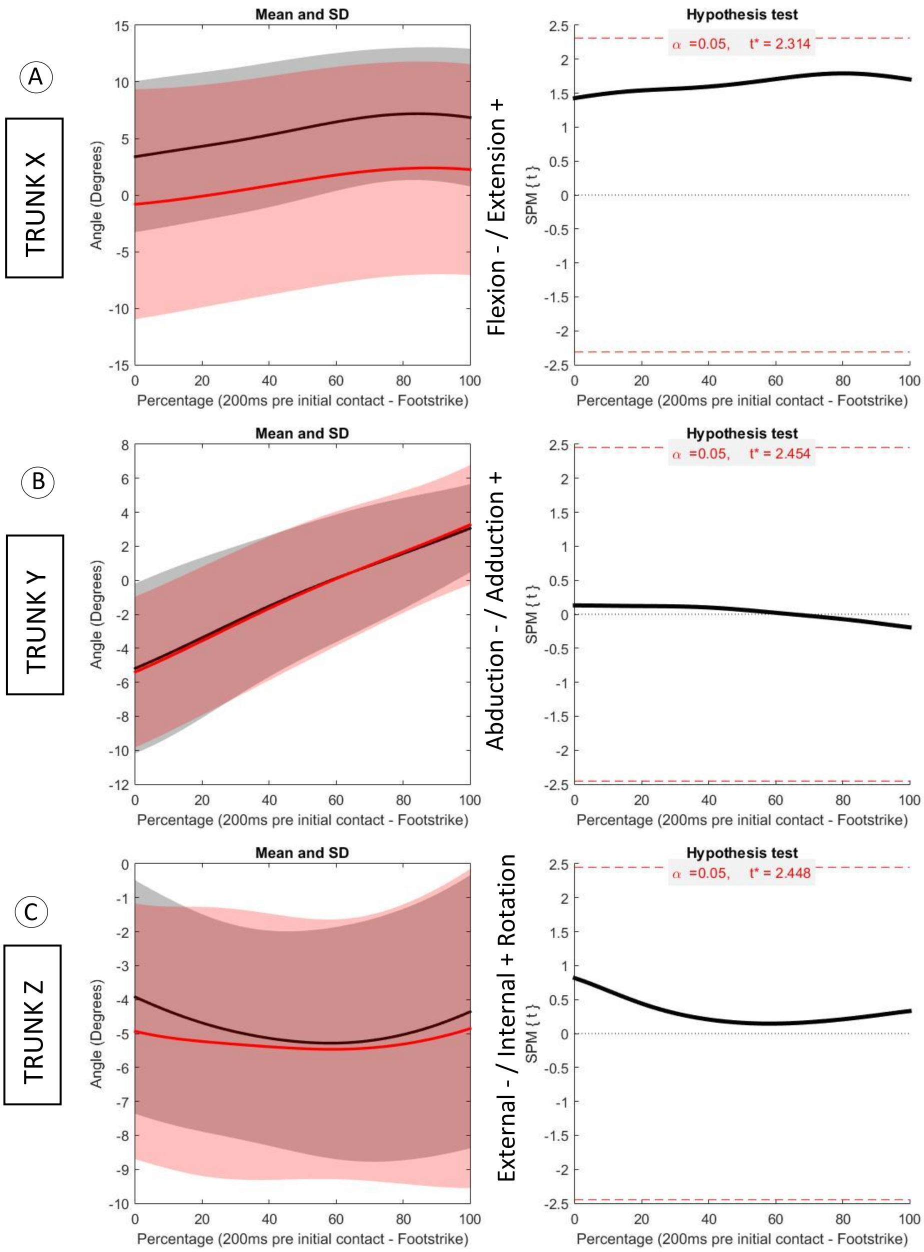


Figure K.5 Cutting- Trunk angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

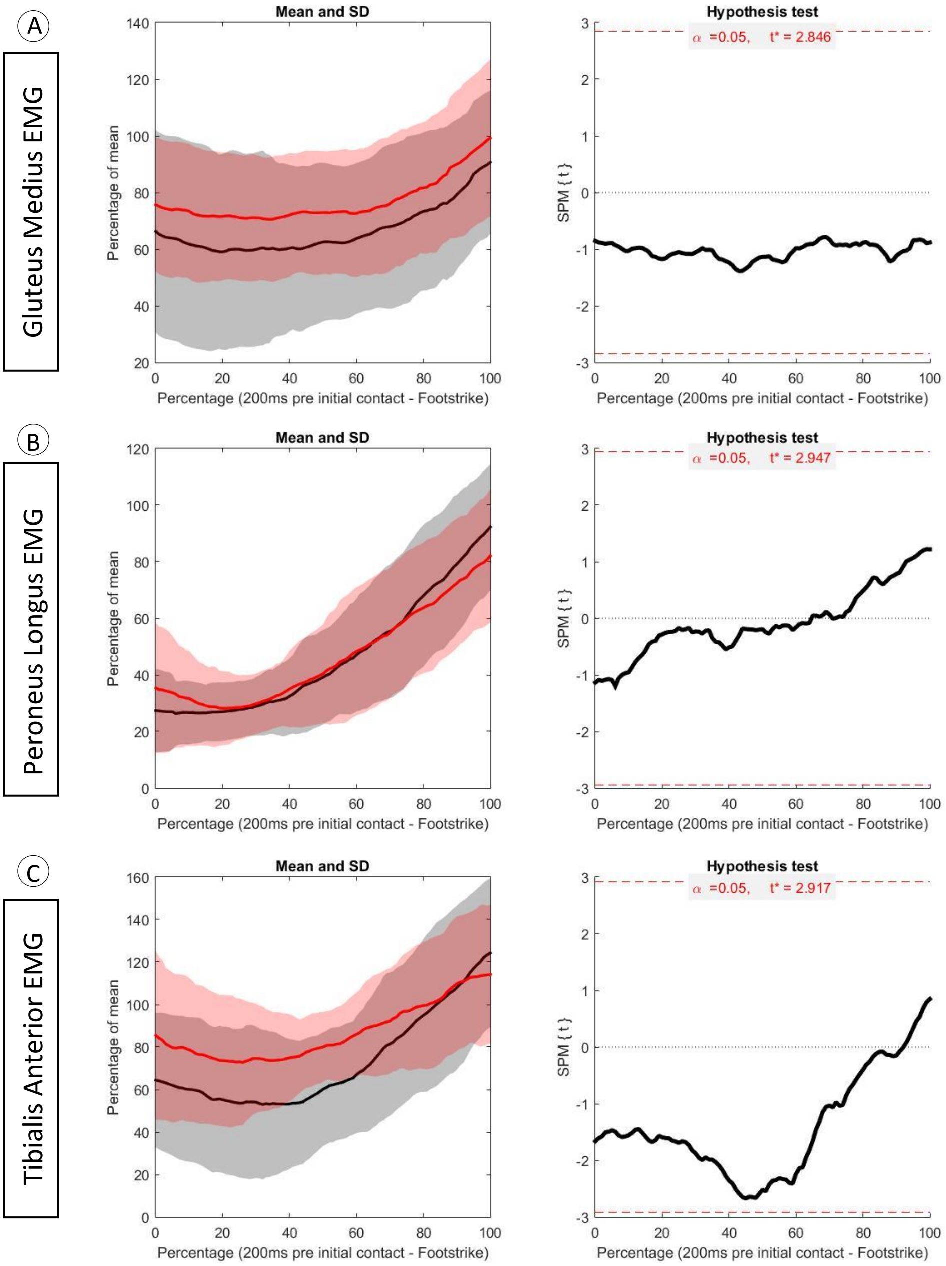


Figure K.6 Cutting- Gluteus Medius and Tibialis Anterior muscle activation - 200 ms pre IC to IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

- **Initial contact to 200 ms post initial contact**

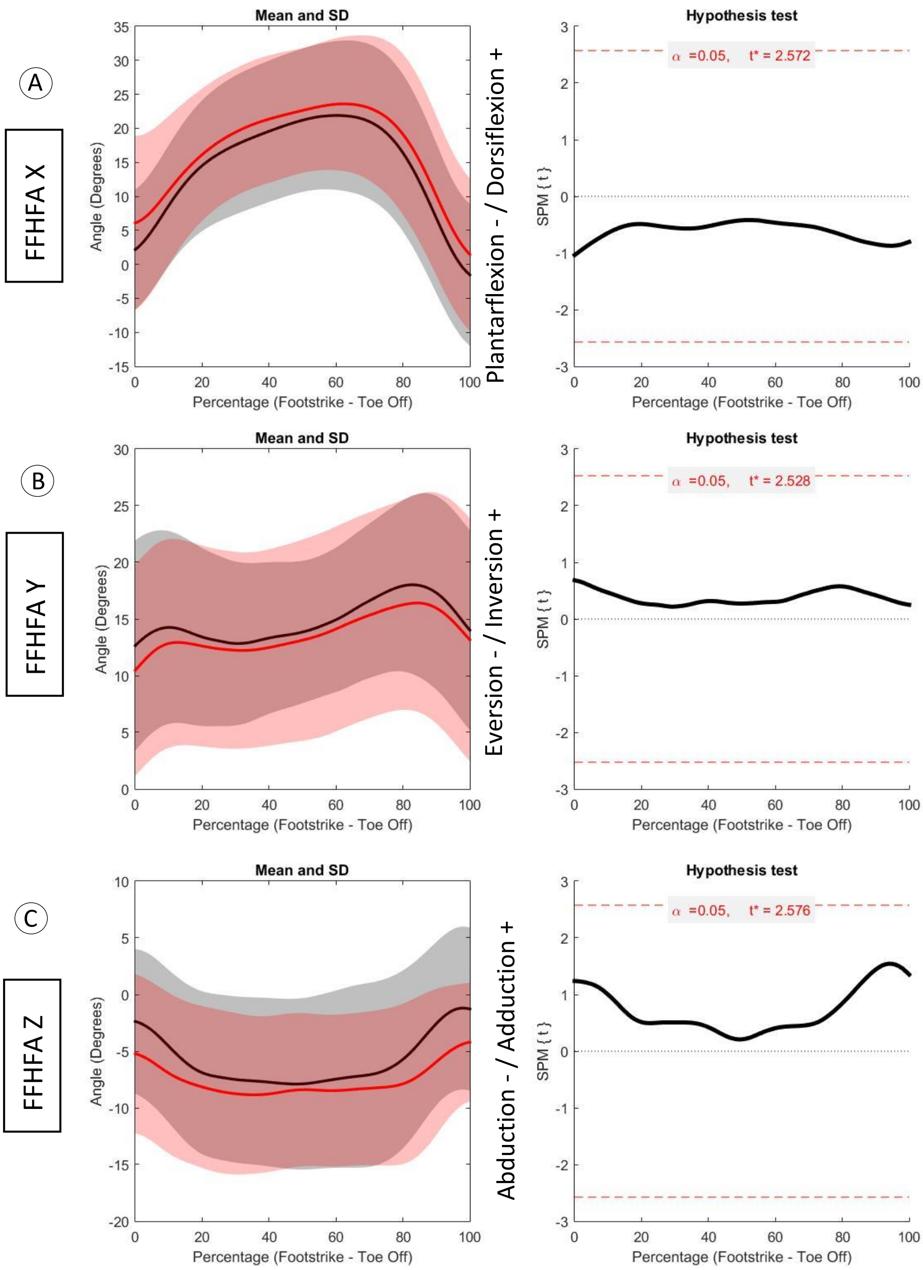


Figure K.7 Cutting- Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

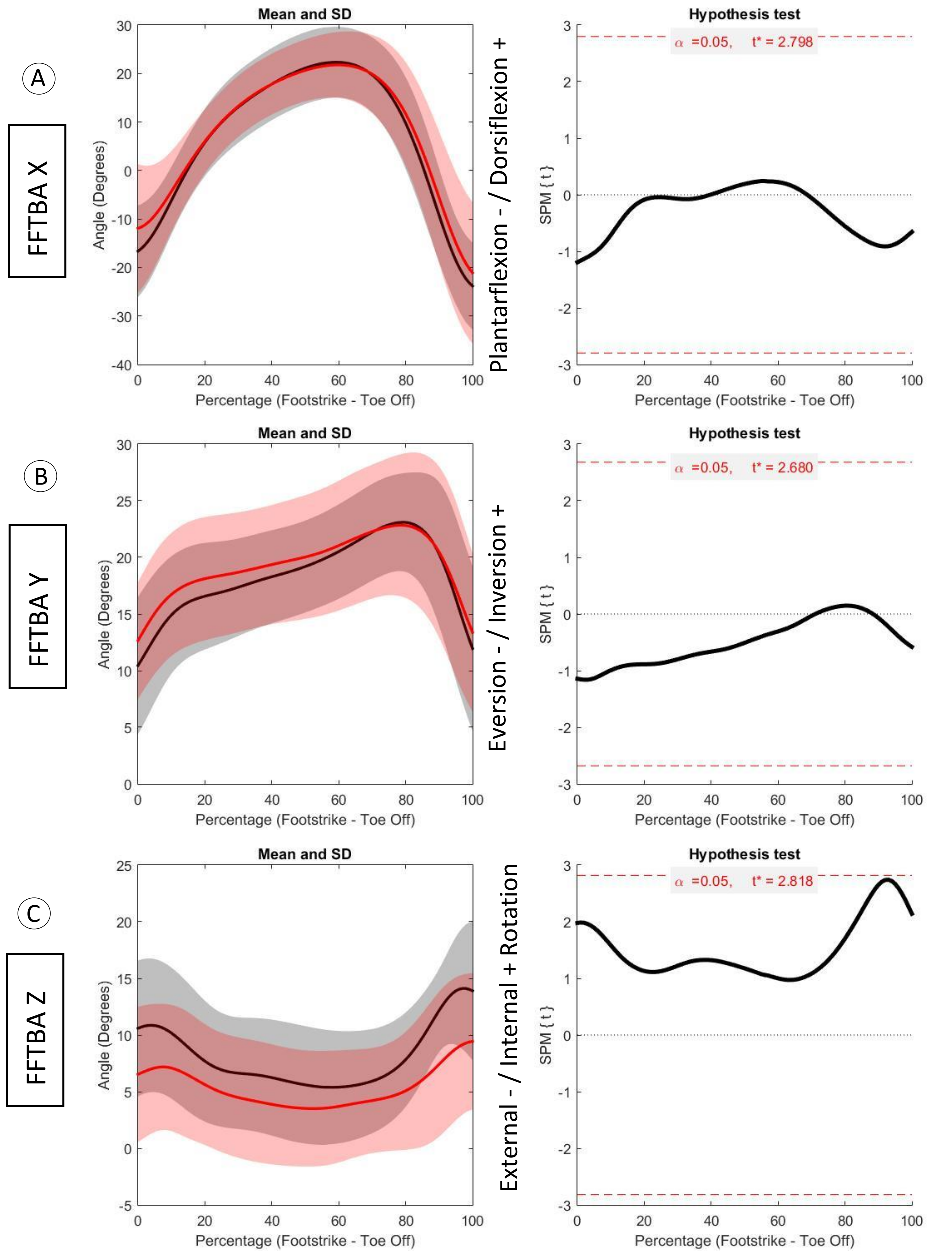


Figure K.8 Cutting- Forefoot –Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

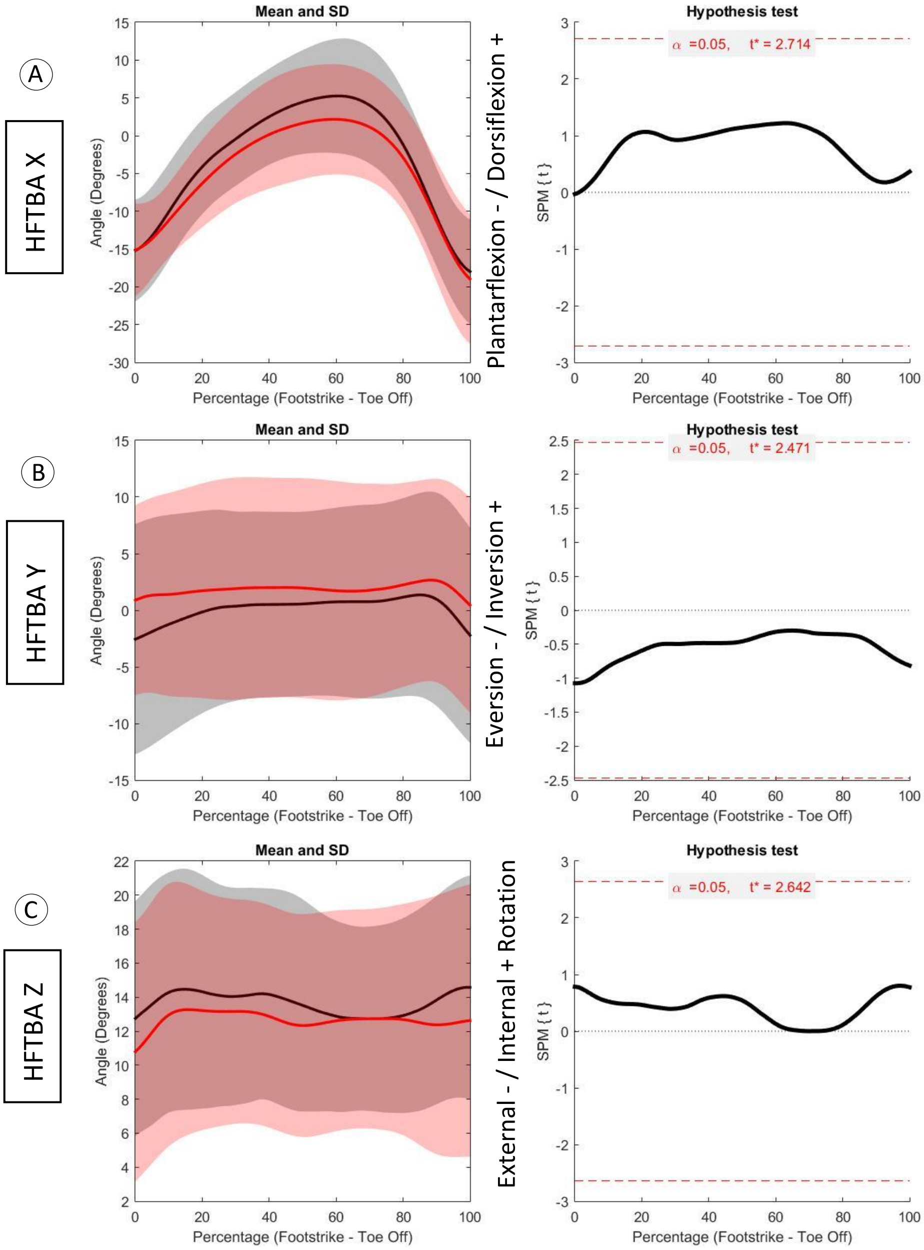


Figure K.9 Cutting- Hindfoot –Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

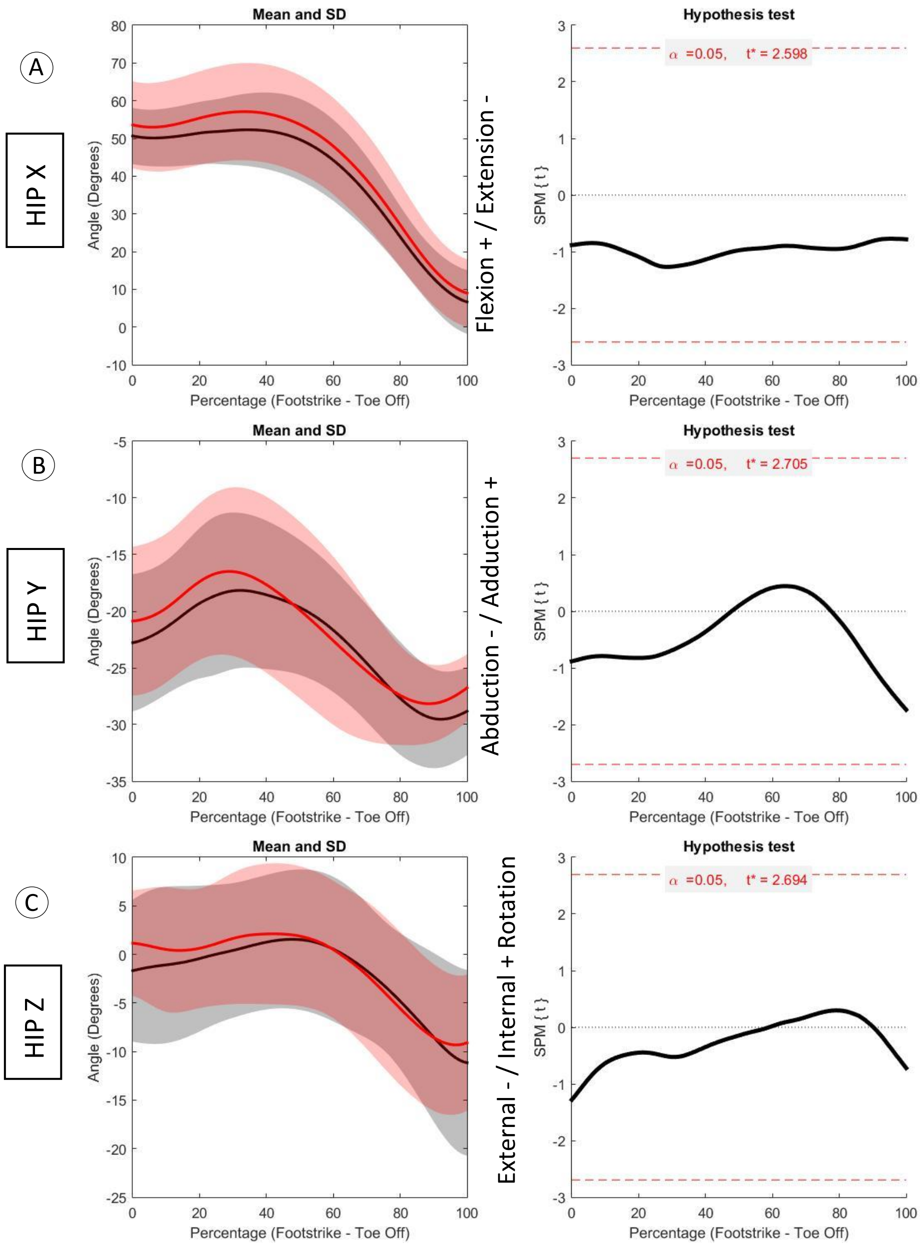


Figure K.10 Cutting- Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

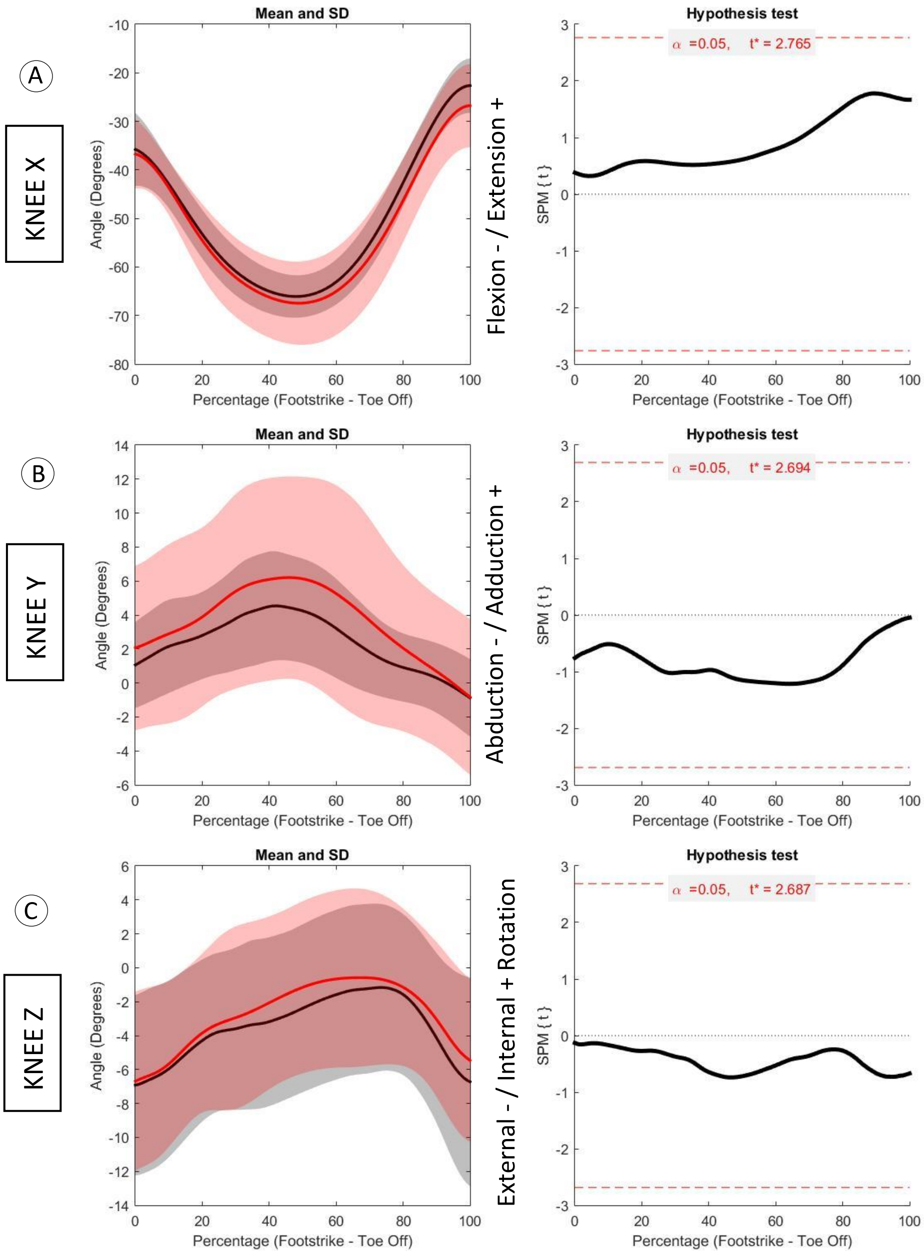


Figure K.11 Cutting- Knee angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

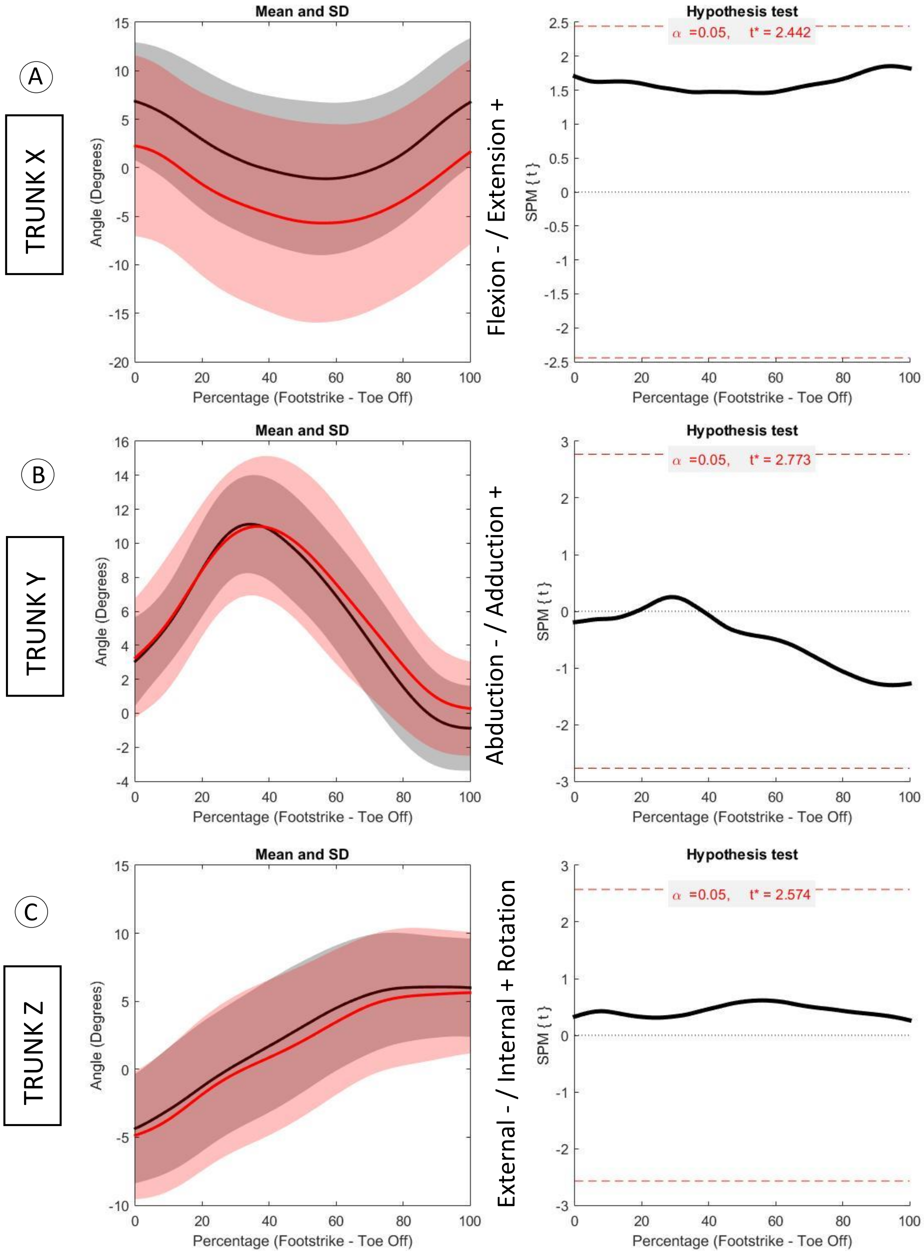


Figure K.12 Cutting- Trunk angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

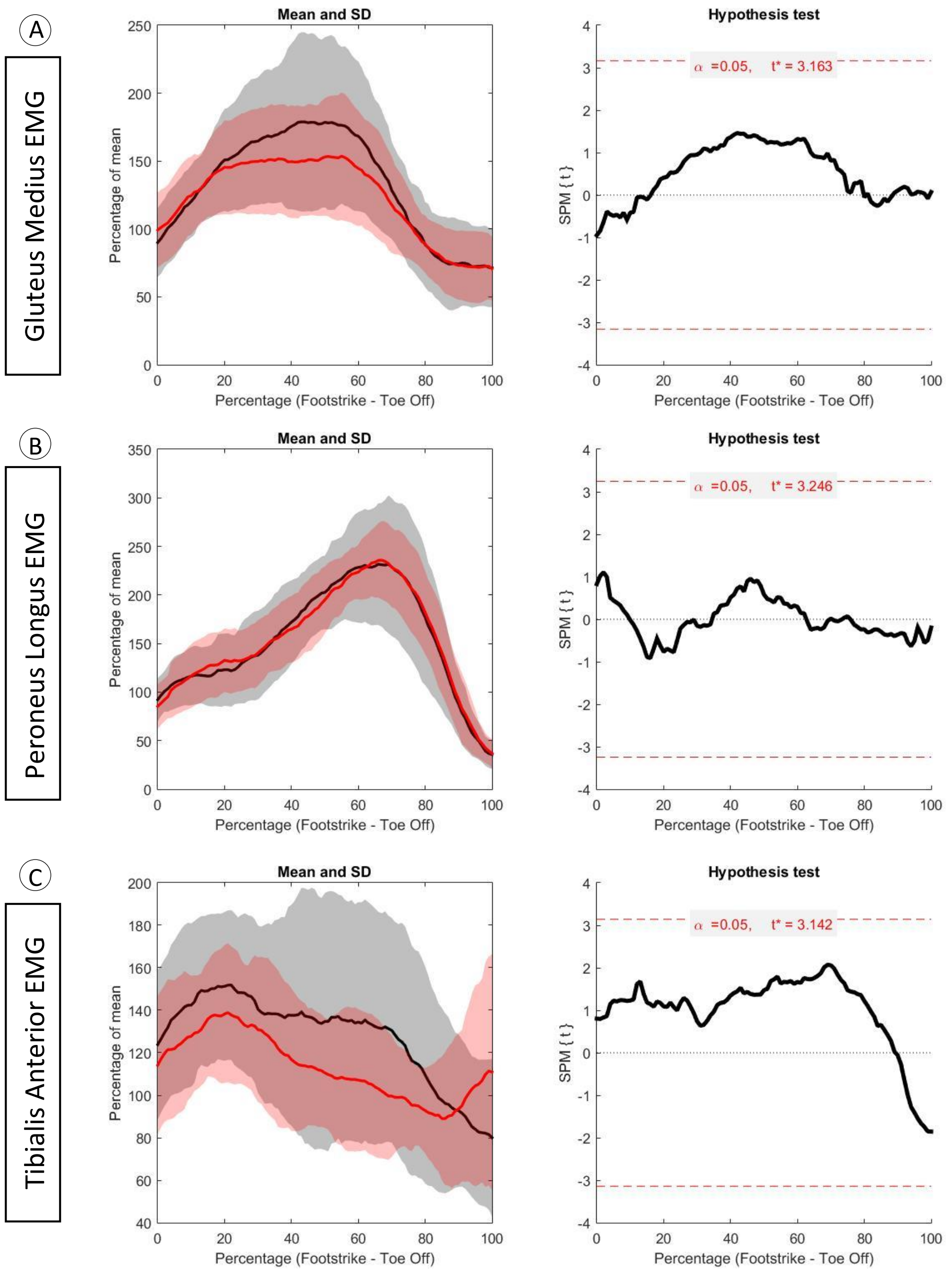


Figure K.13 Cutting- Gluteus Medius and Tibialis Anterior muscle activation – IC to 200 ms post IC - means and standard deviations (Matched Control - — Affected - —) and *t*-test output.

Unaffected limb vs Affected limb of CAI group

- **200 ms pre initial contact to initial contact**

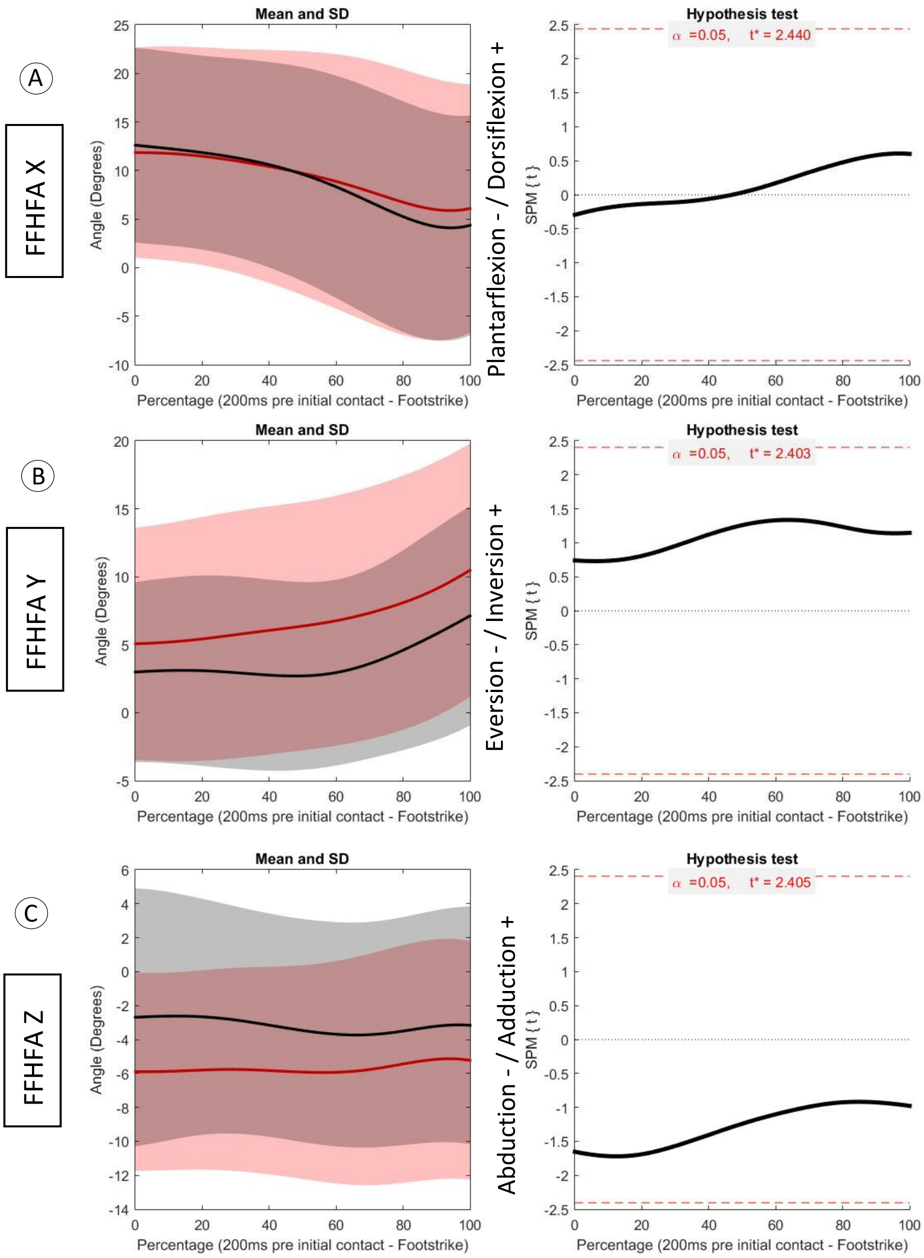


Figure K.14 Cutting- Forefoot – Hindfoot angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

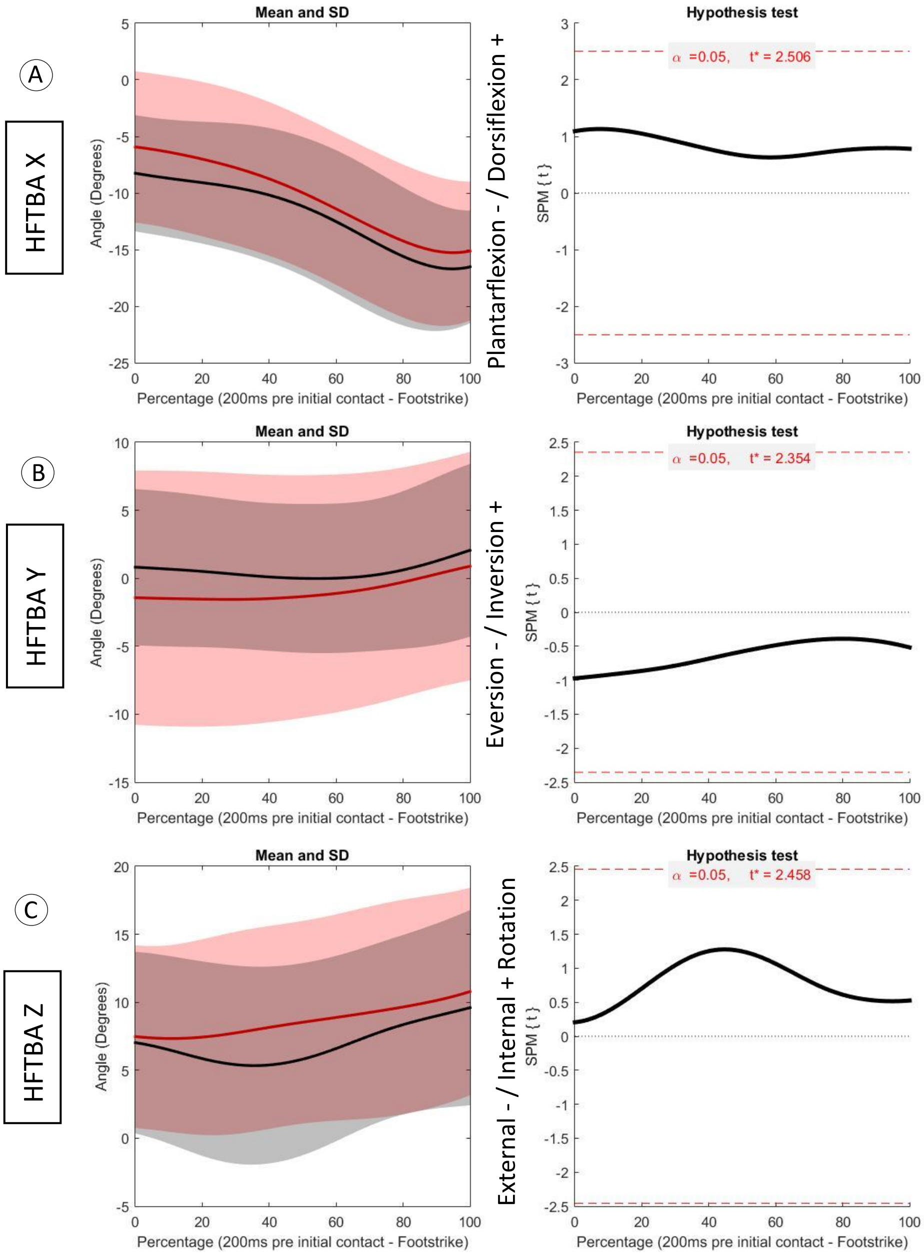


Figure K.15 Cutting- Hindfoot – Tibia angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

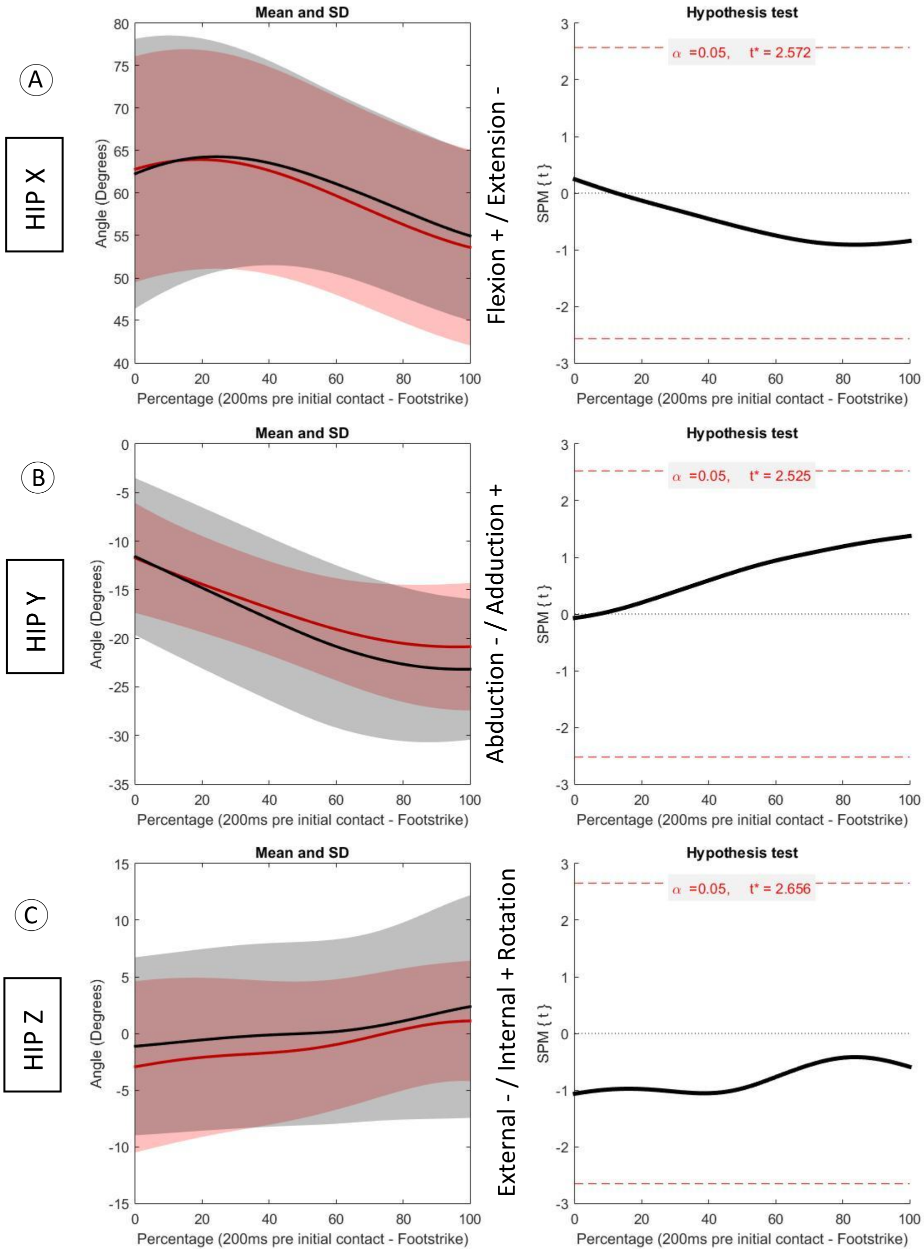


Figure K.16 Cutting- Hip angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

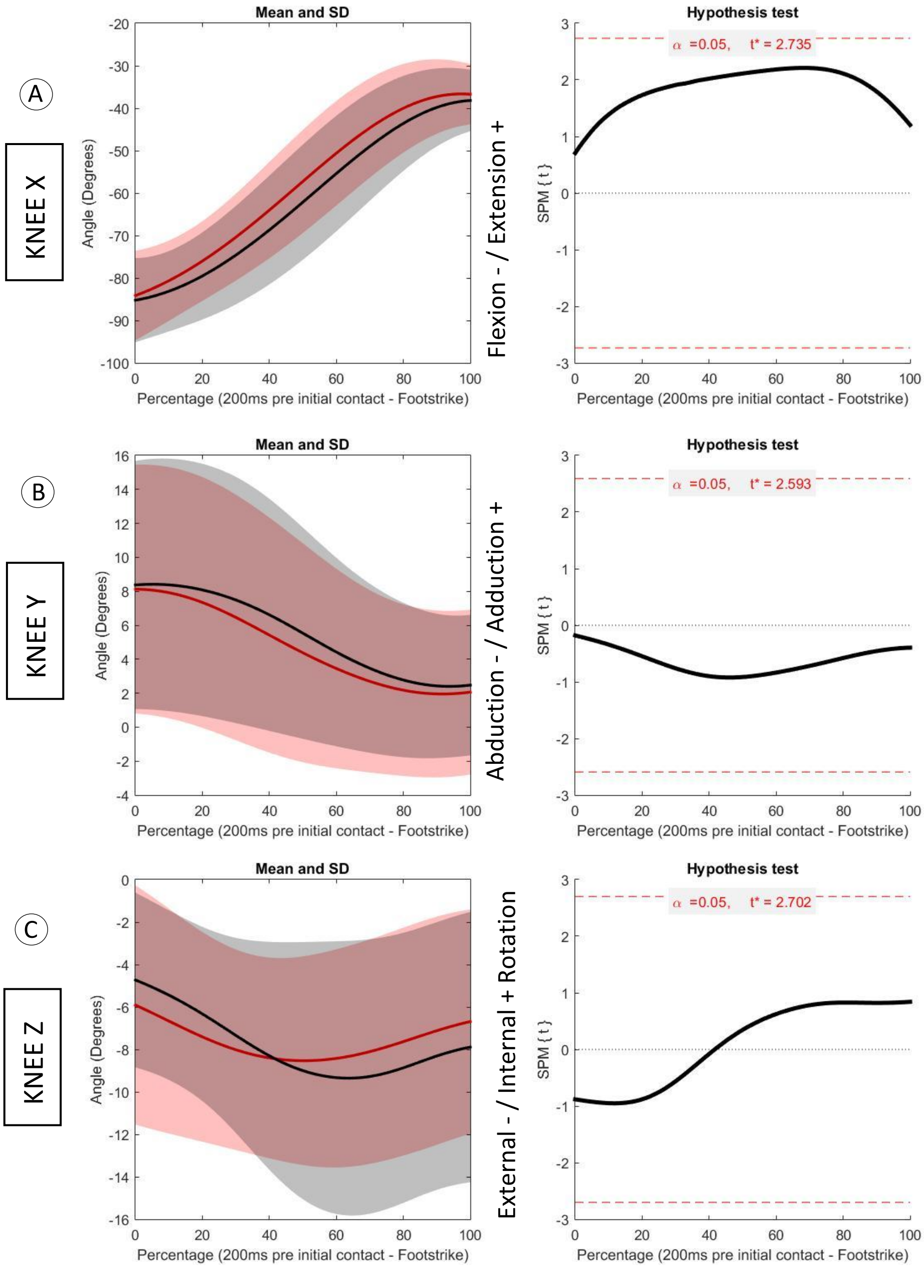


Figure K.17 Cutting- Knee angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

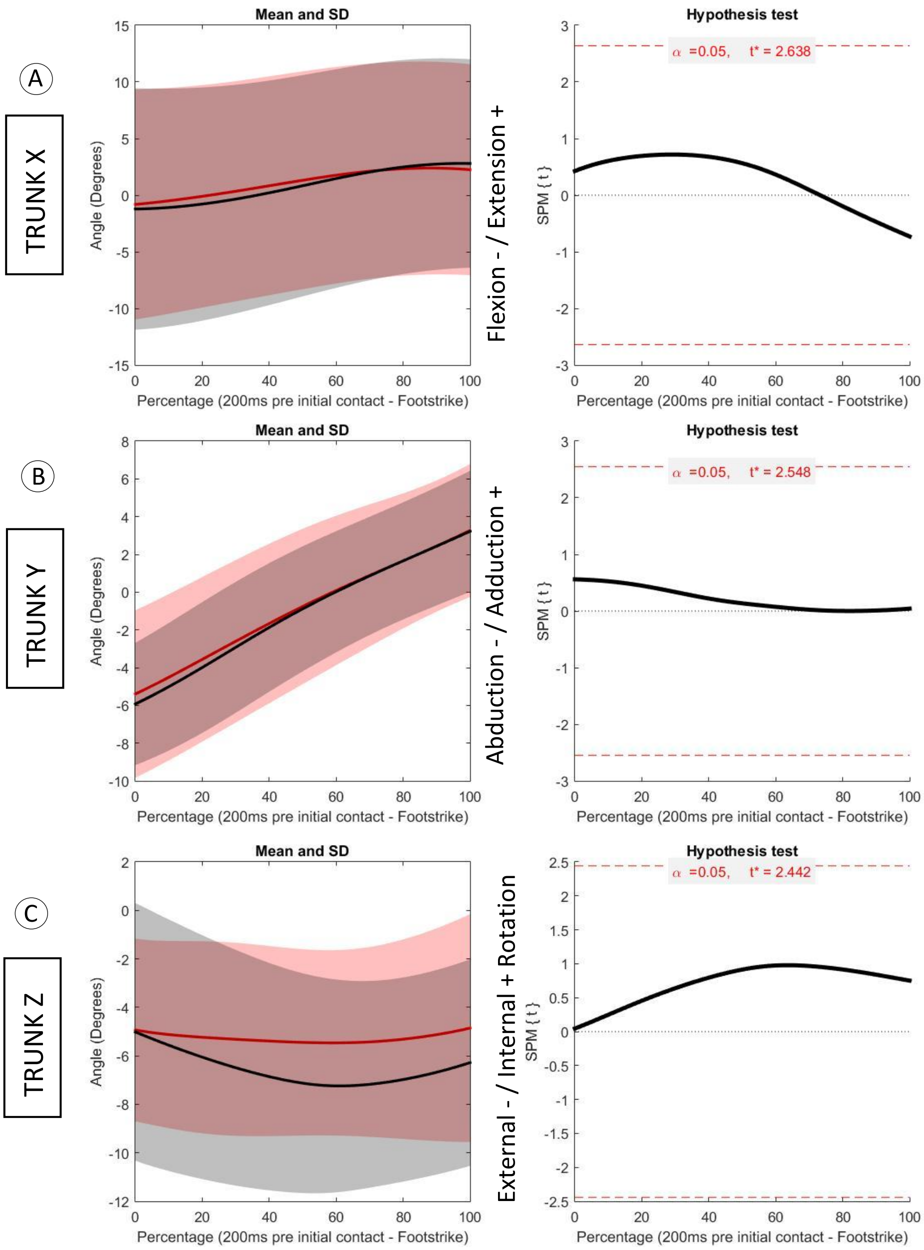


Figure K.18 Cutting- Trunk angles (x, y, z) 200 ms pre IC to IC – means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

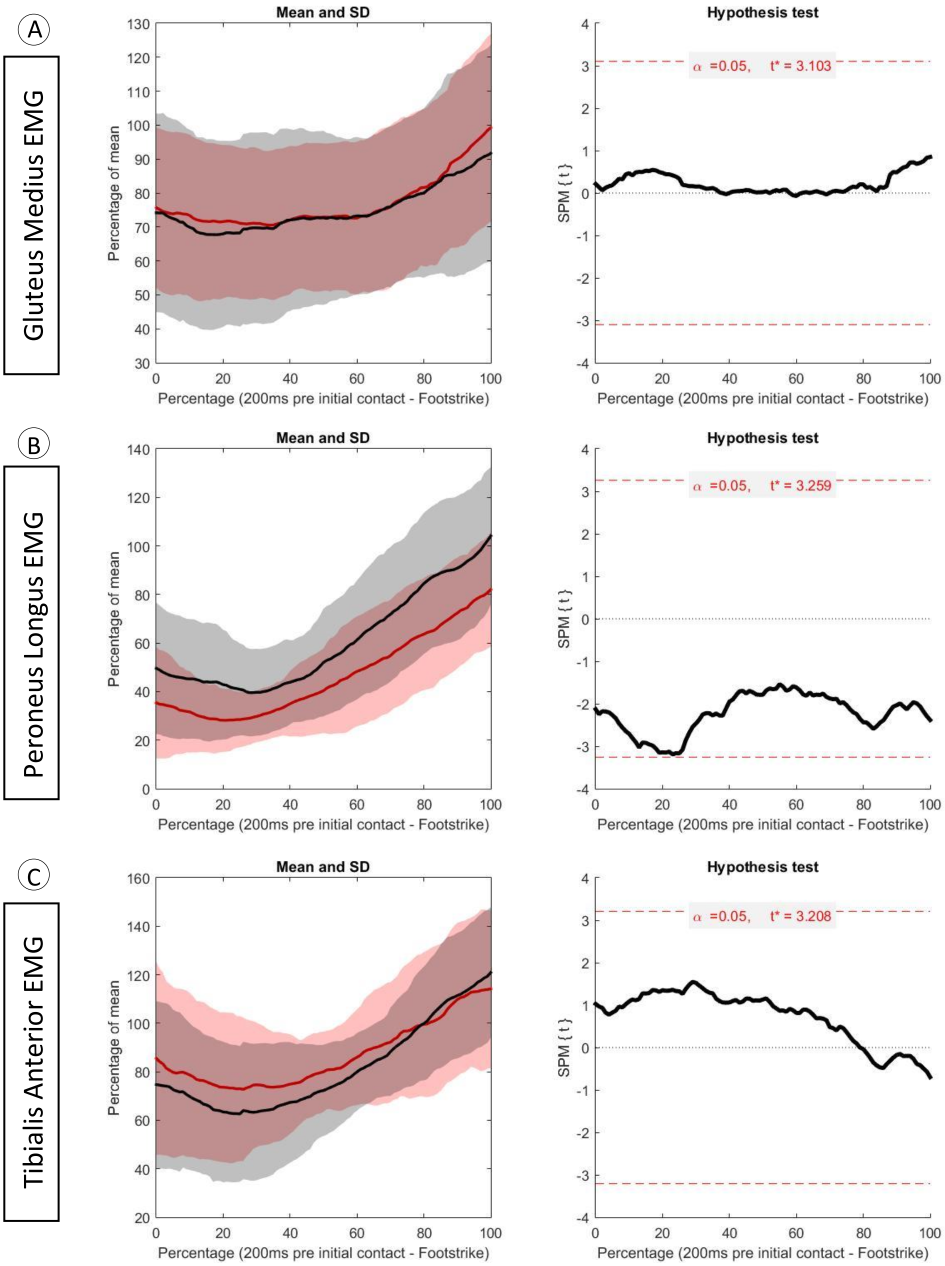


Figure K.19 Cutting- Gluteus Medius and Tibialis Anterior muscle activation – 200 ms pre IC to IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

- **Initial contact to 200 ms pre initial contact**

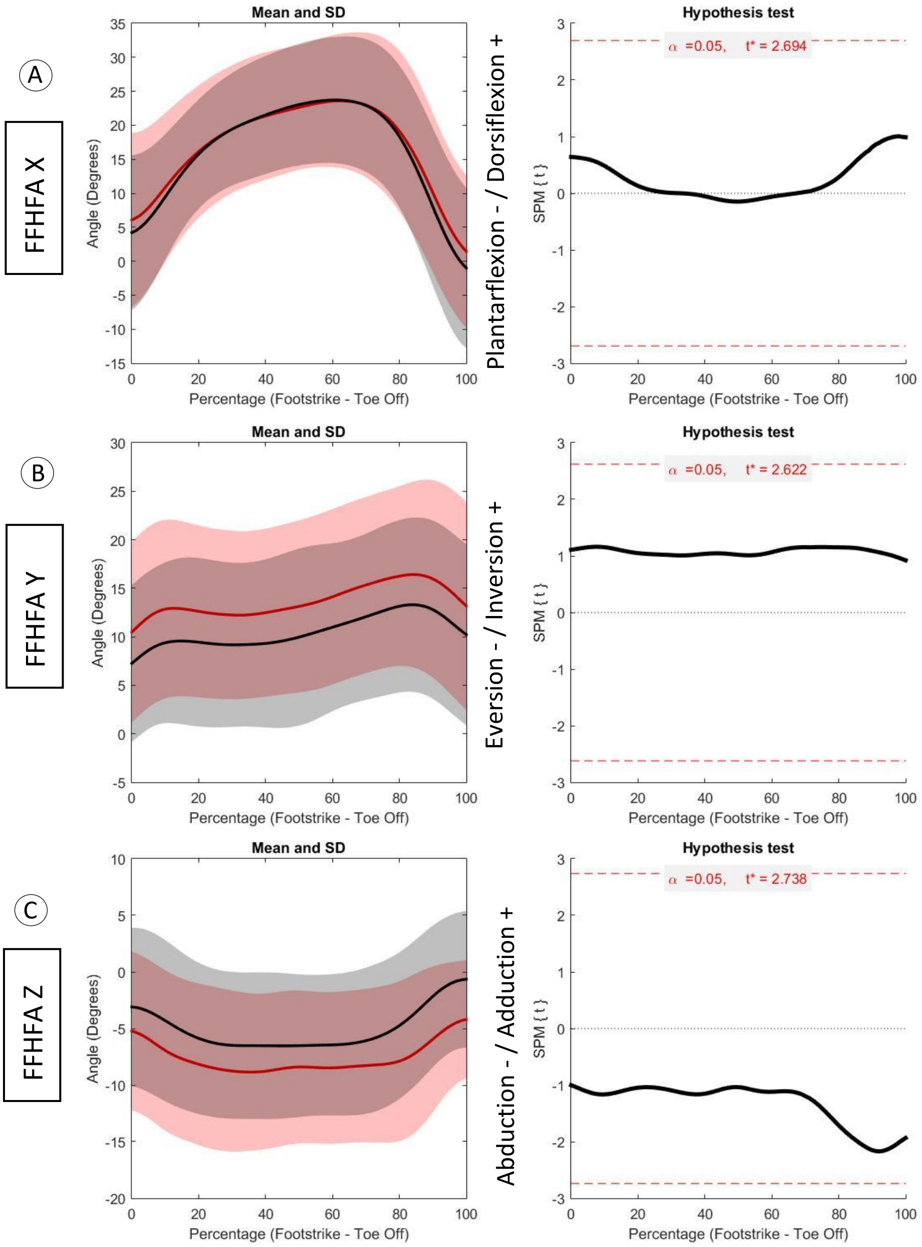


Figure K.20 Cutting- Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

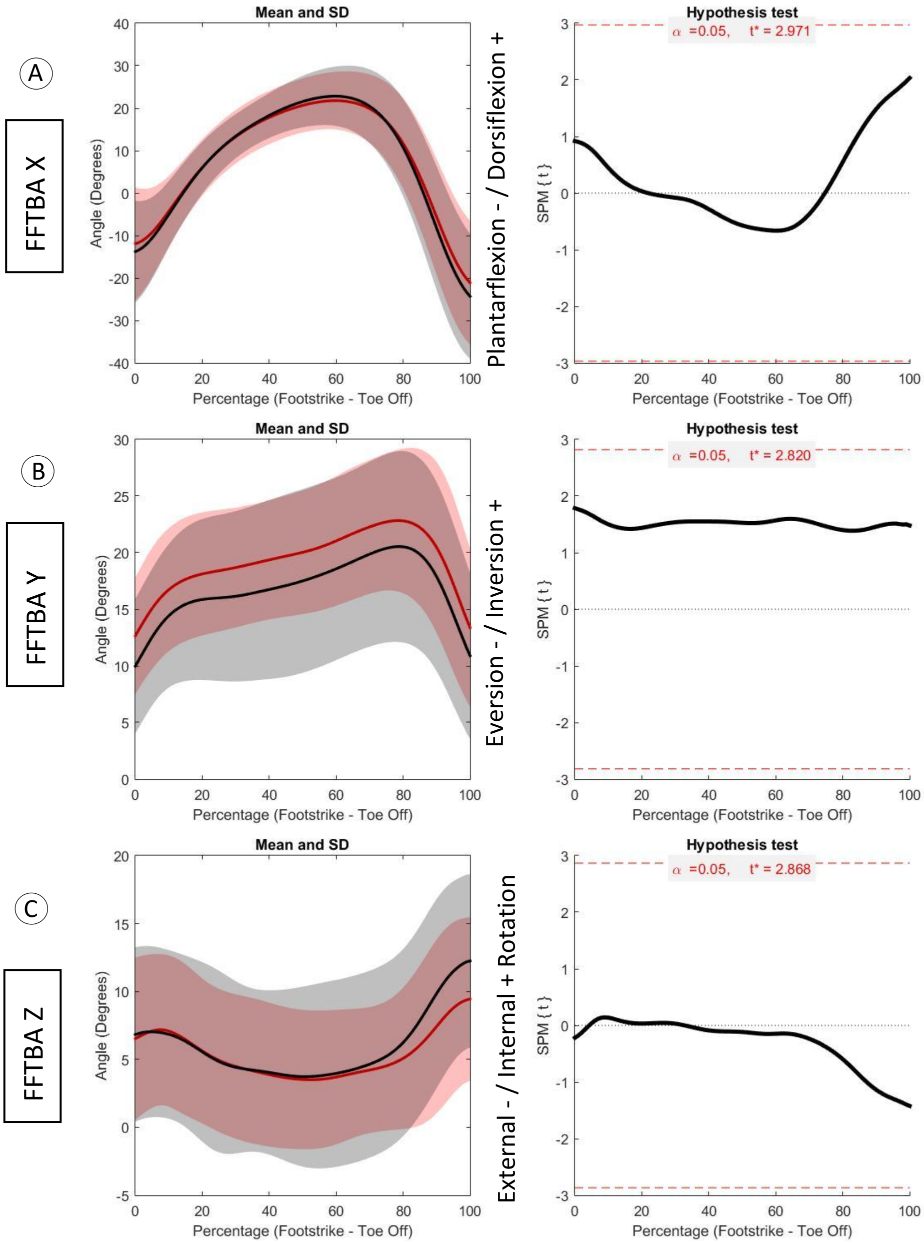


Figure K.21 Cutting- Forefoot – Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

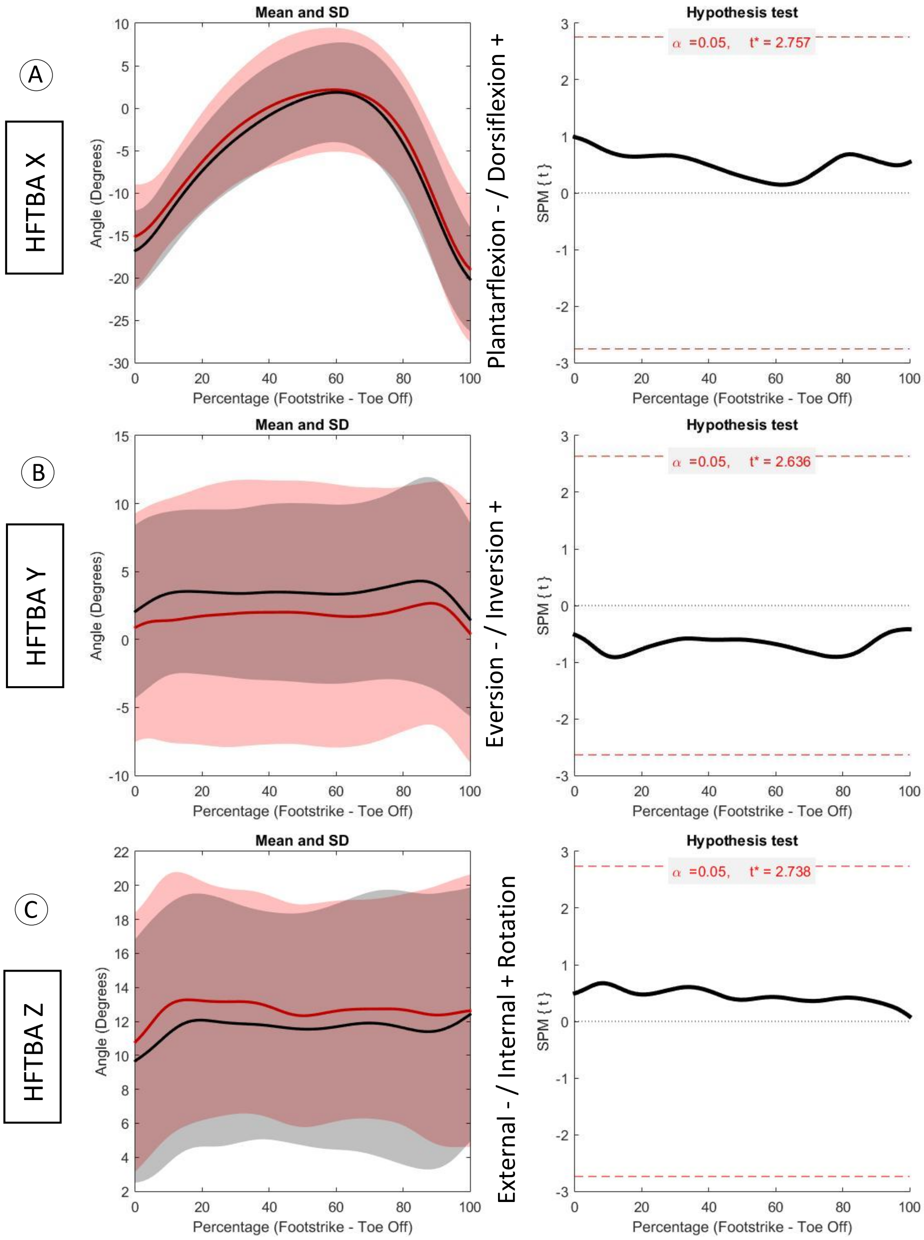


Figure K.22 Cutting- Hindfoot – Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

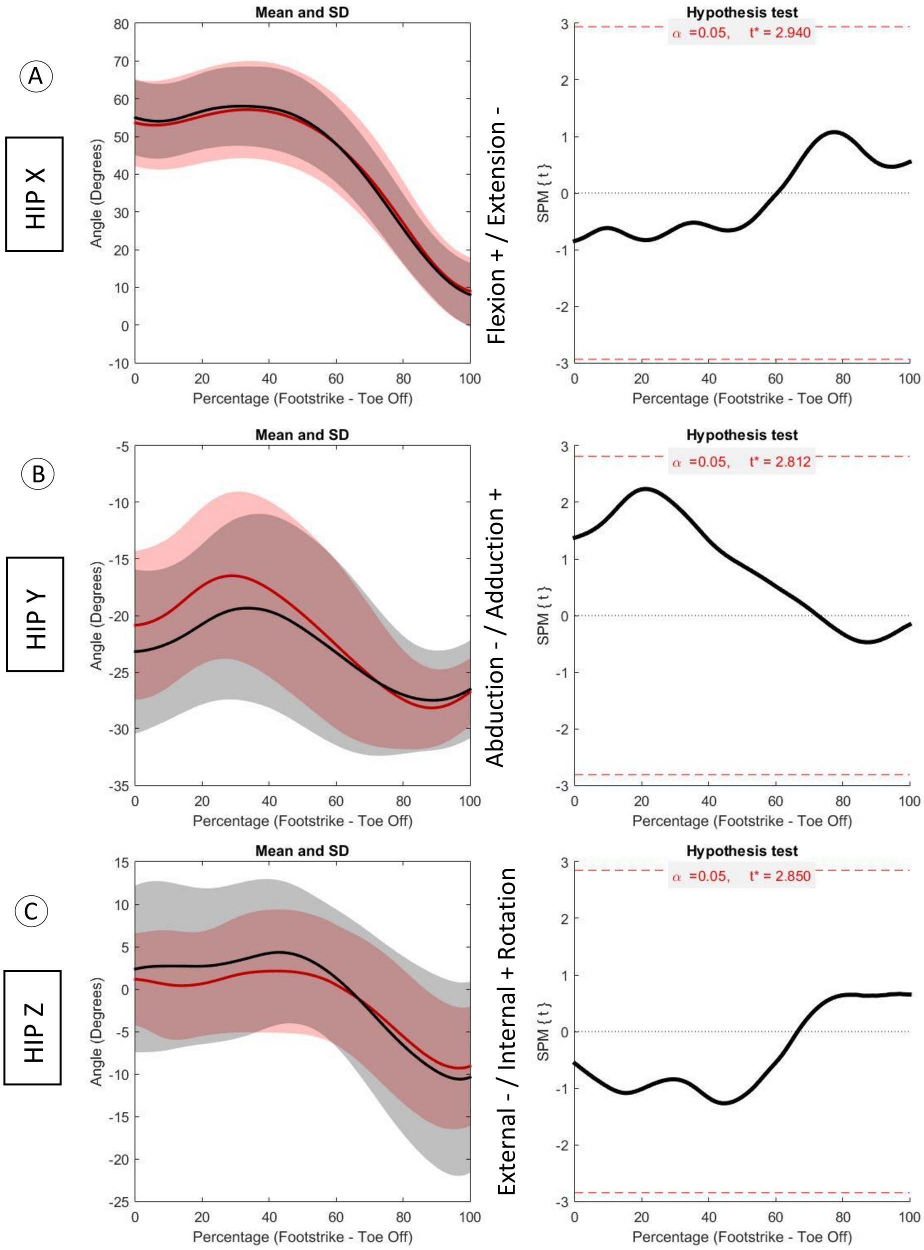


Figure K.23 Cutting- Hip angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

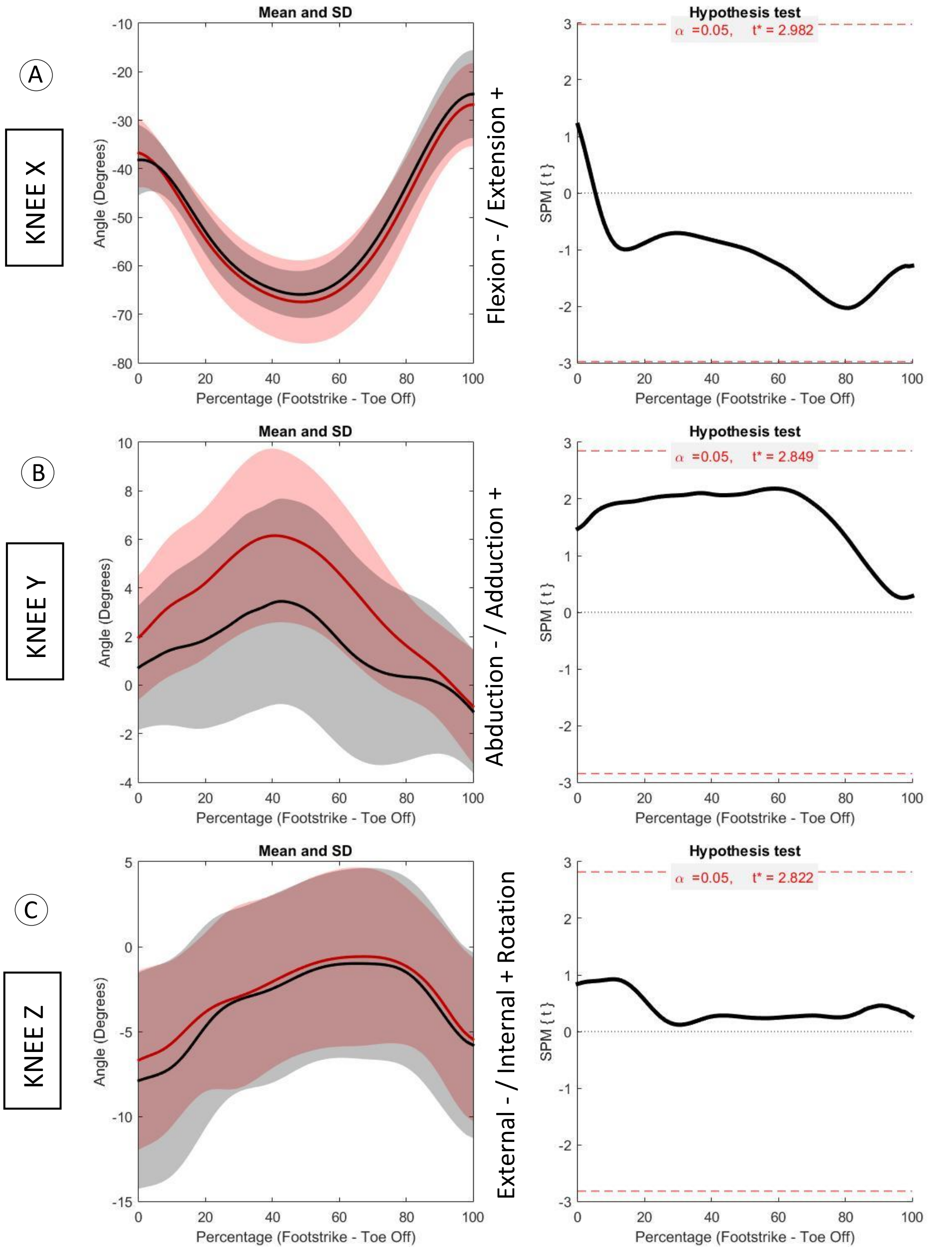


Figure K.24 Cutting- Knee angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

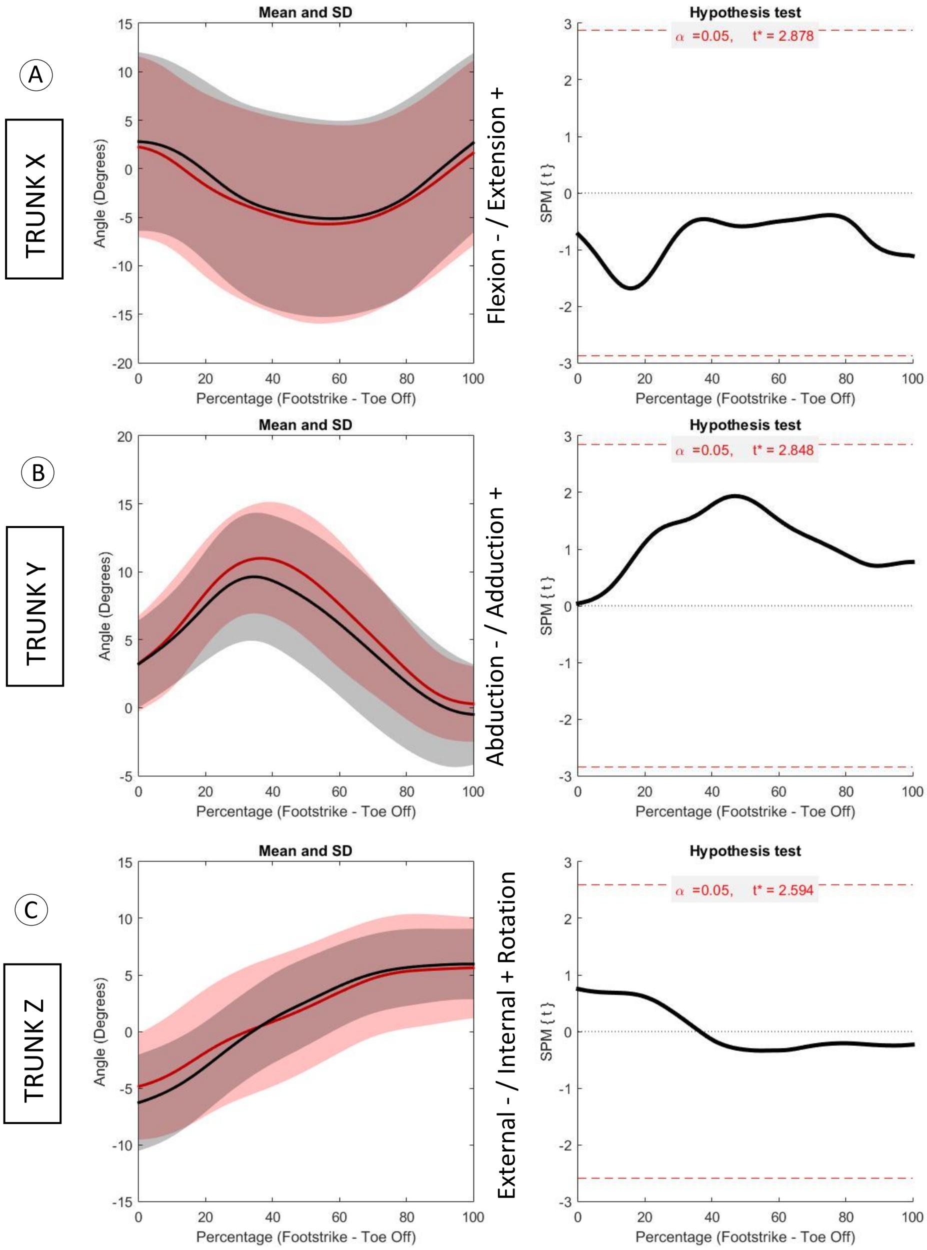


Figure K.25 Cutting- Trunk angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

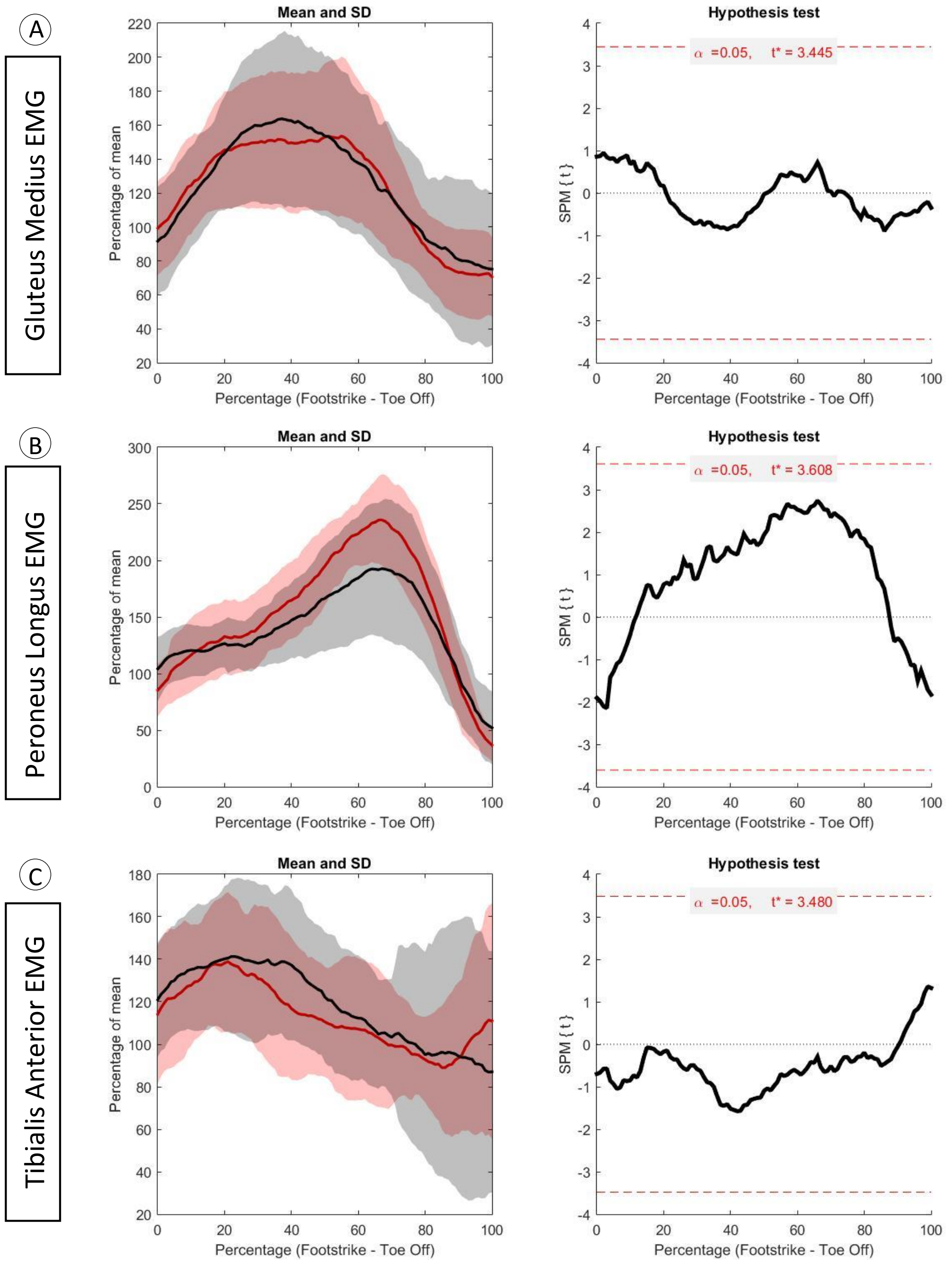


Figure K.26 Cutting- Gluteus Medius and Tibialis Anterior muscle activation – IC to 200 ms post IC - means and standard deviations (Unaffected - — Affected - —) and *t*-test output.

Matched Control vs Unaffected limb of CAI group

- **200 ms pre initial contact to initial contact**

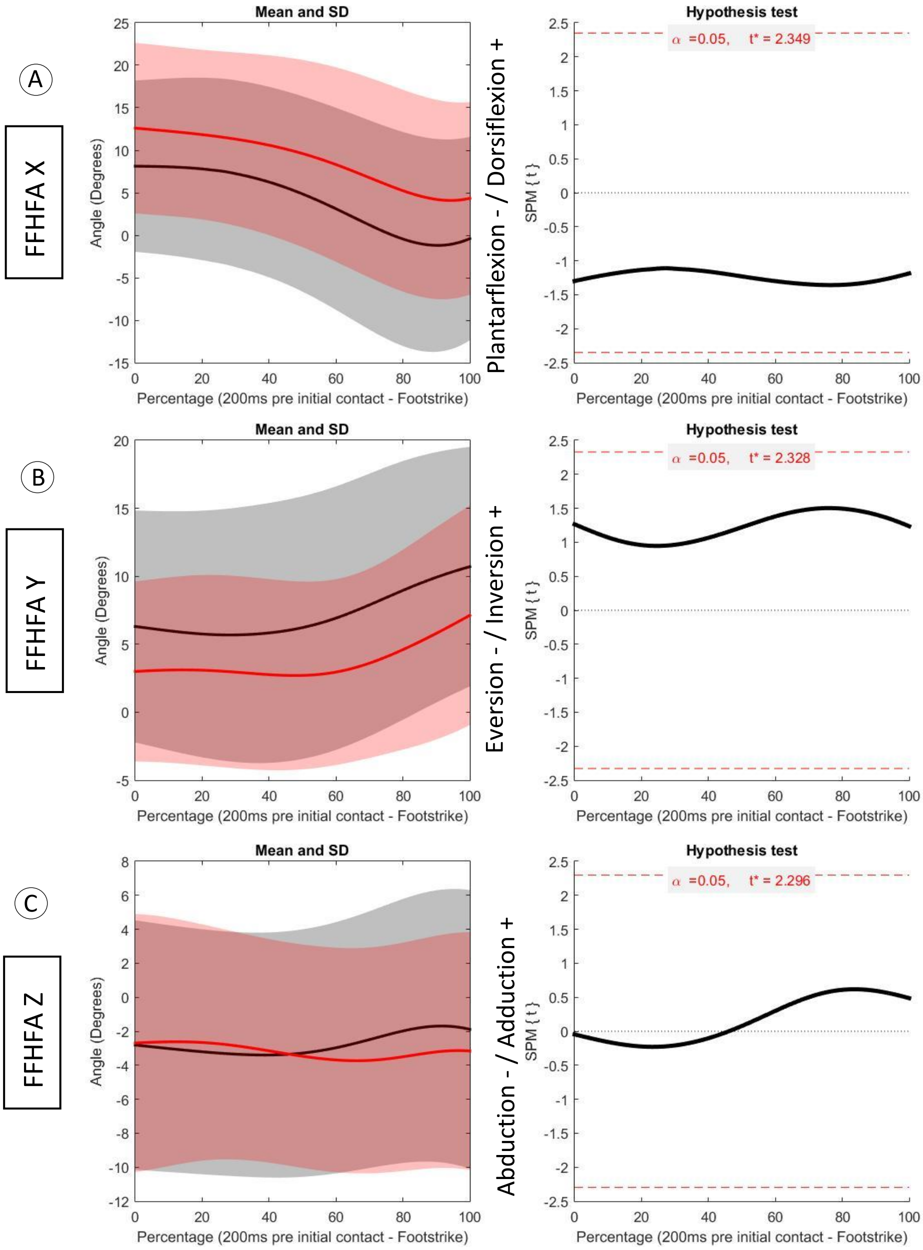


Figure K.27 Cutting- Forefoot – Hindfoot angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

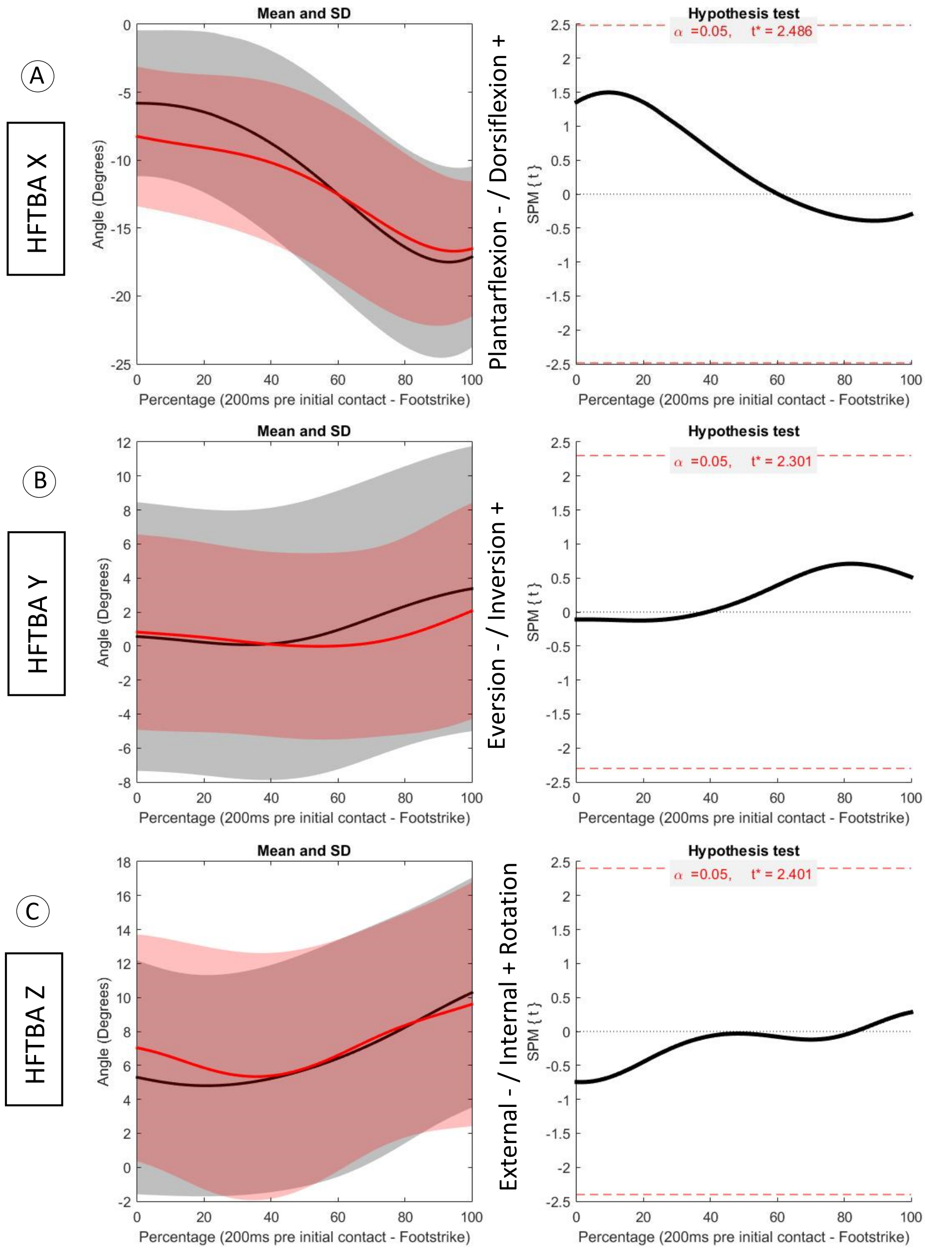


Figure K.28 Cutting- Hindfoot – tibia angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

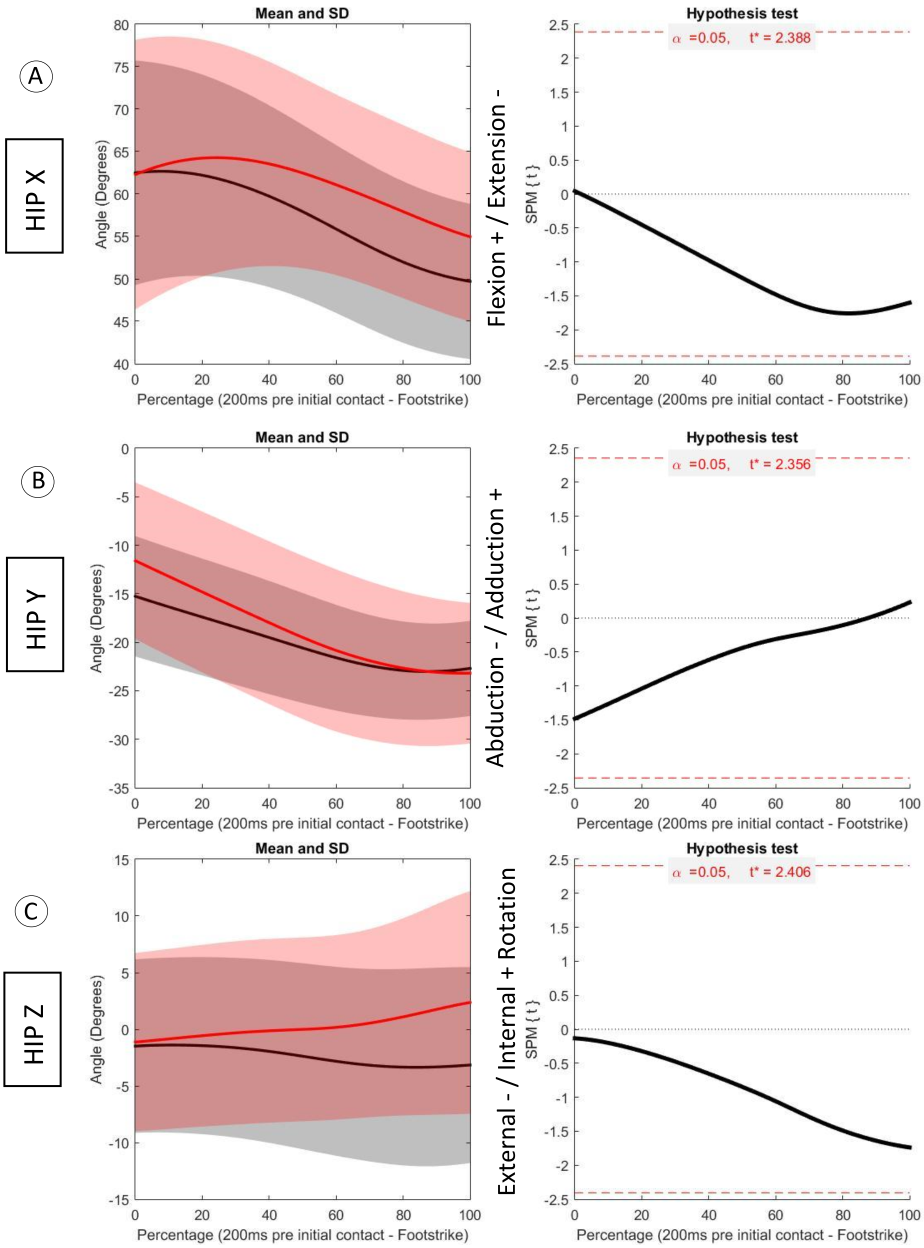


Figure K.29 Cutting- Hip angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

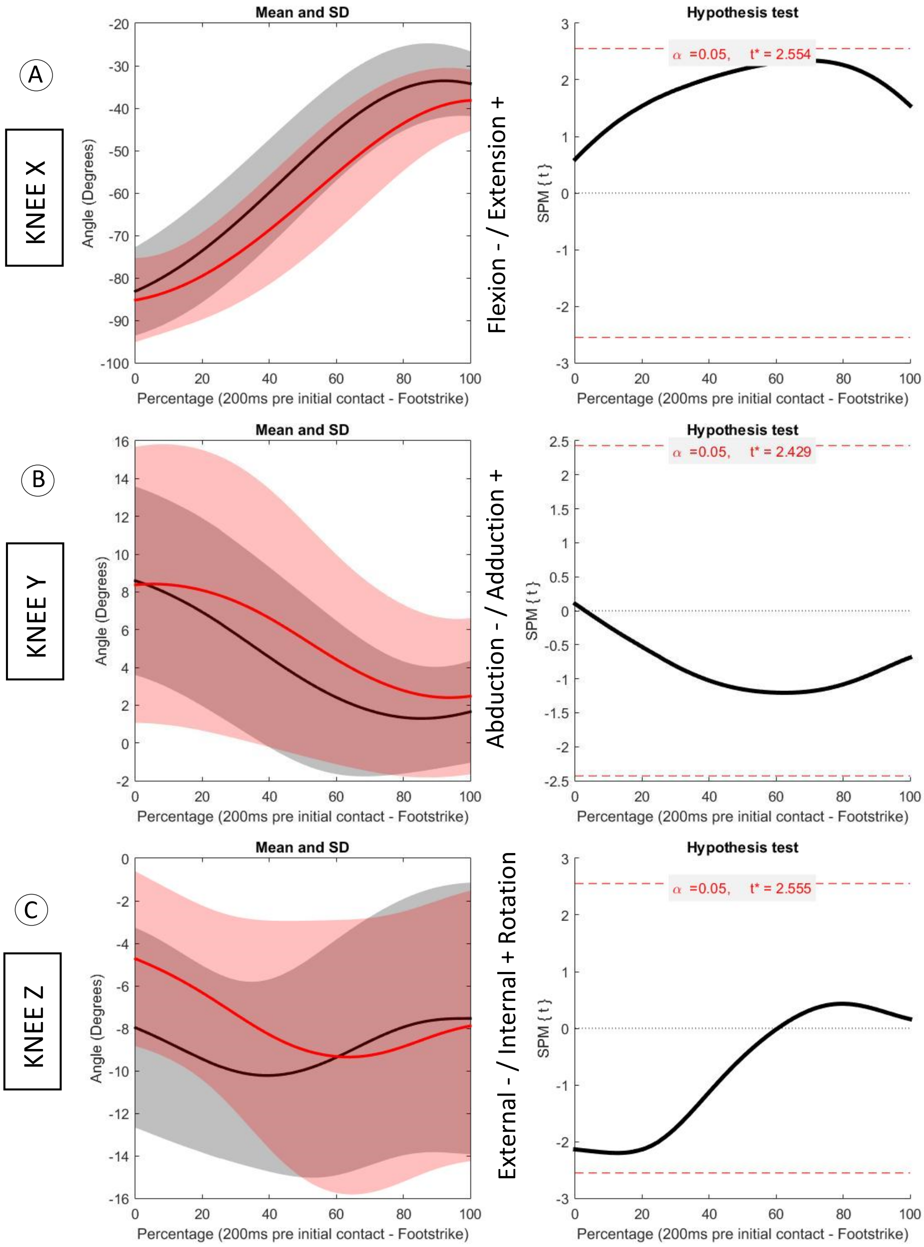


Figure K.30 Cutting- Knee angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

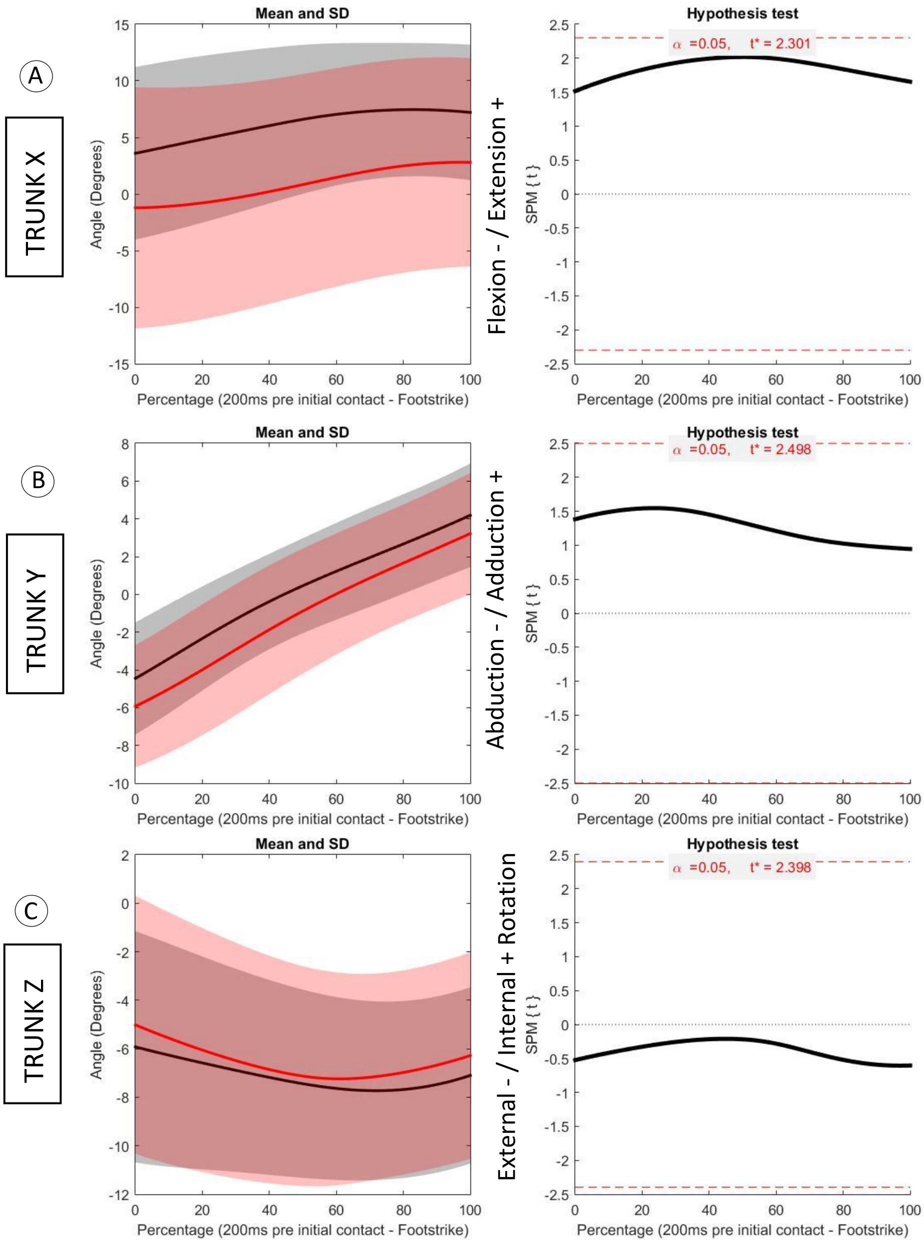


Figure K.31 Cutting- Trunk angles (x, y, z) 200 ms pre IC to IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

Initial contact to 200 ms pre initial contact

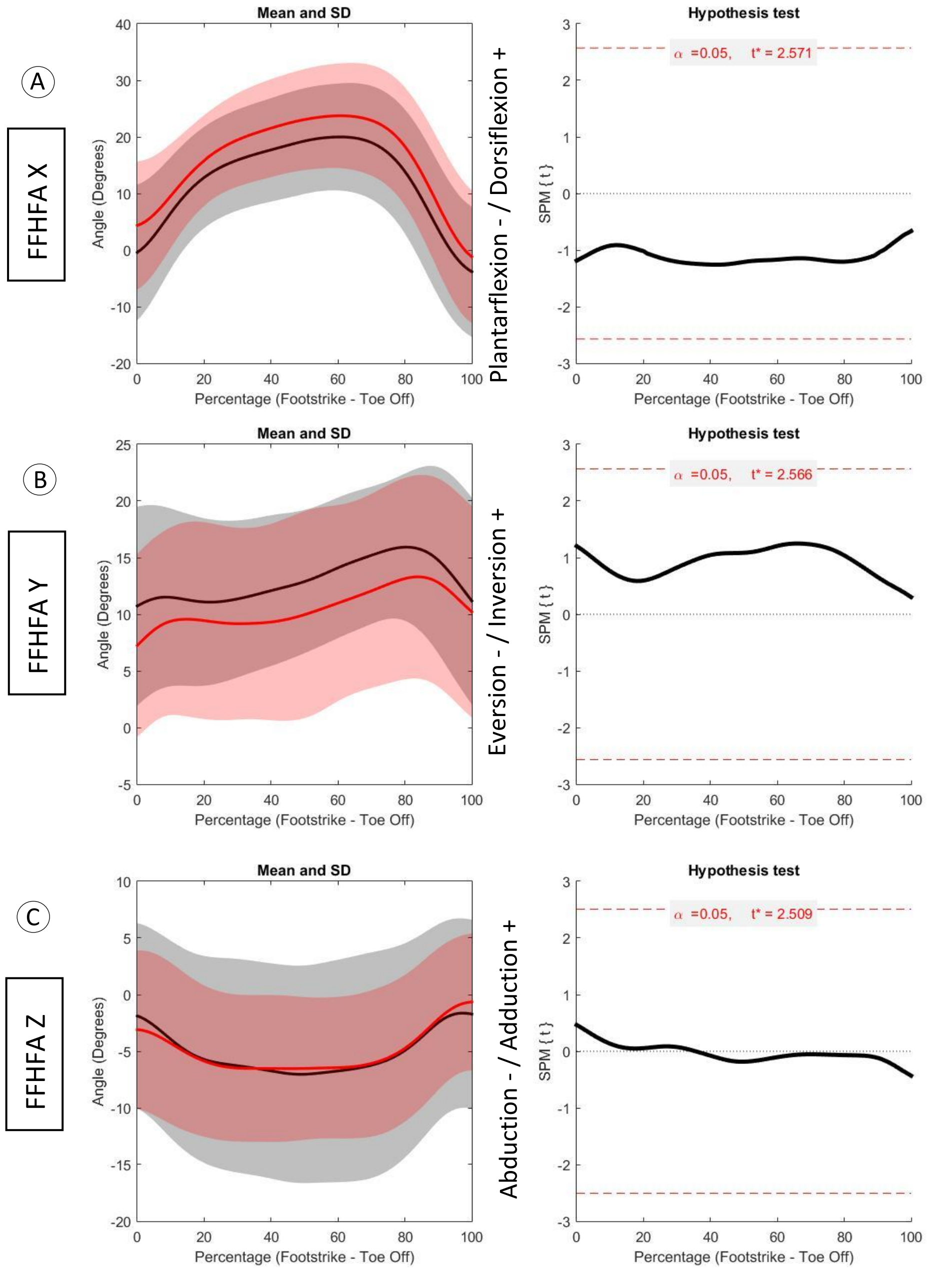


Figure K.32 Cutting- Forefoot – Hindfoot angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

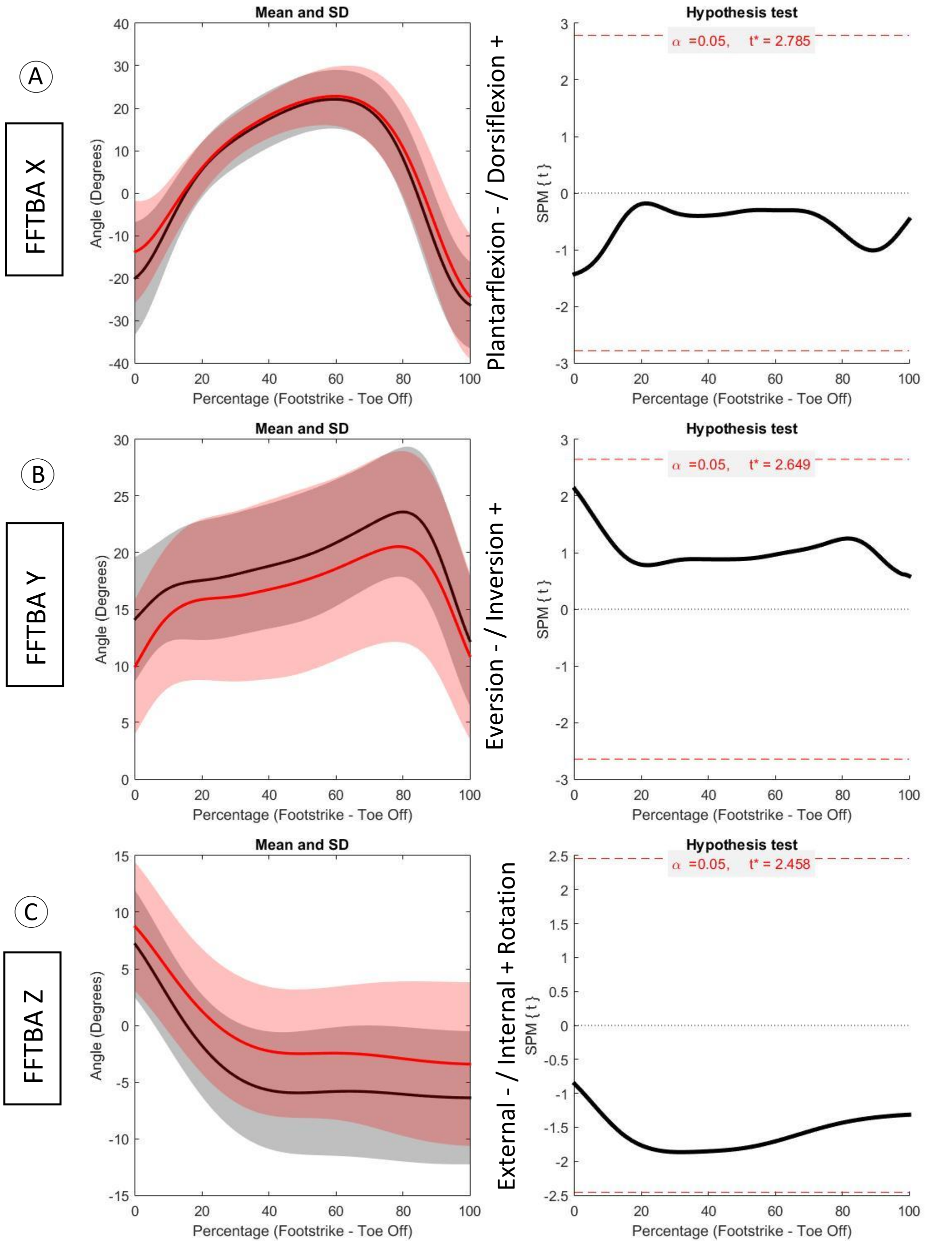


Figure K.33 Cutting- Forefoot –Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

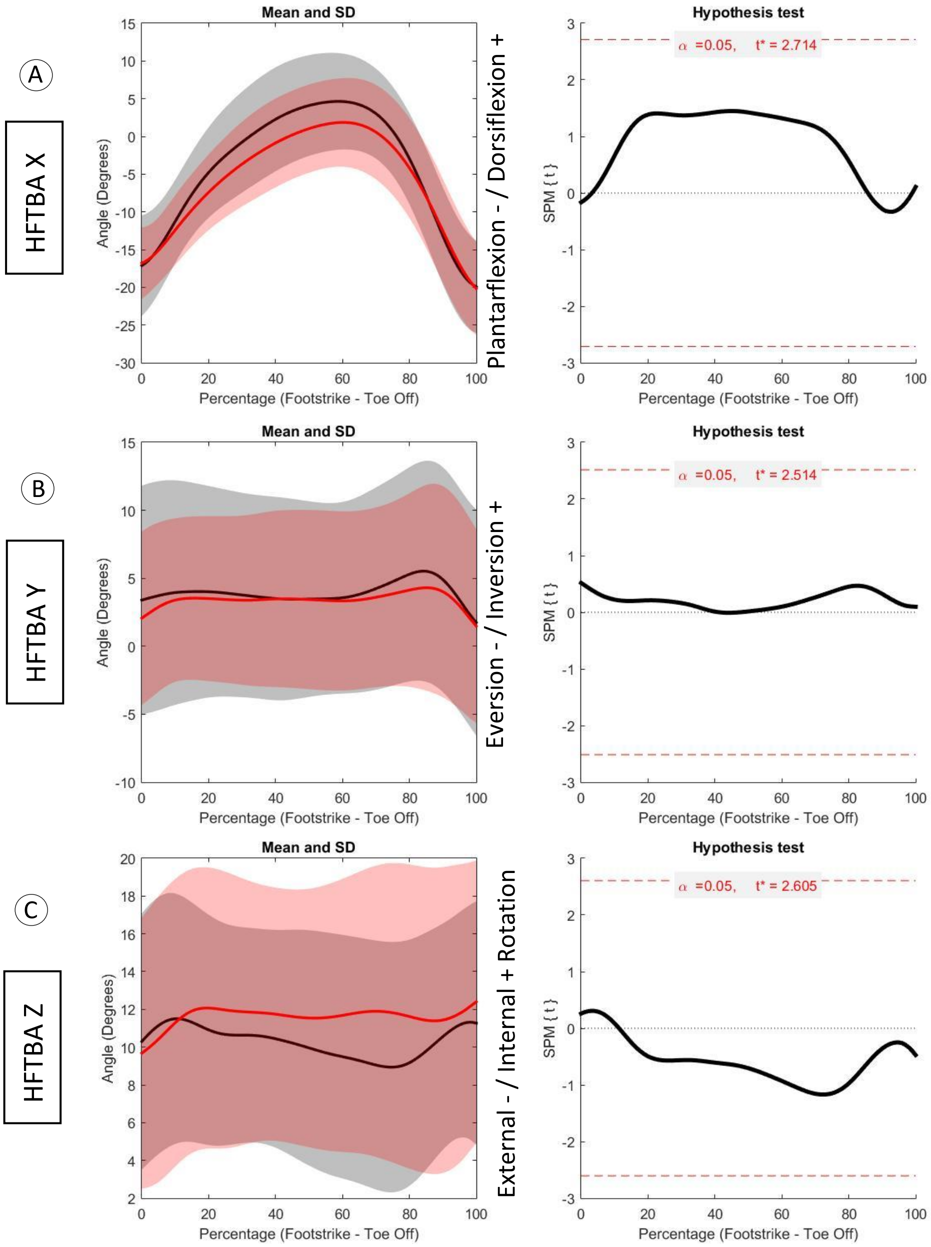


Figure K.34 Cutting- Hindfoot-Tibia angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

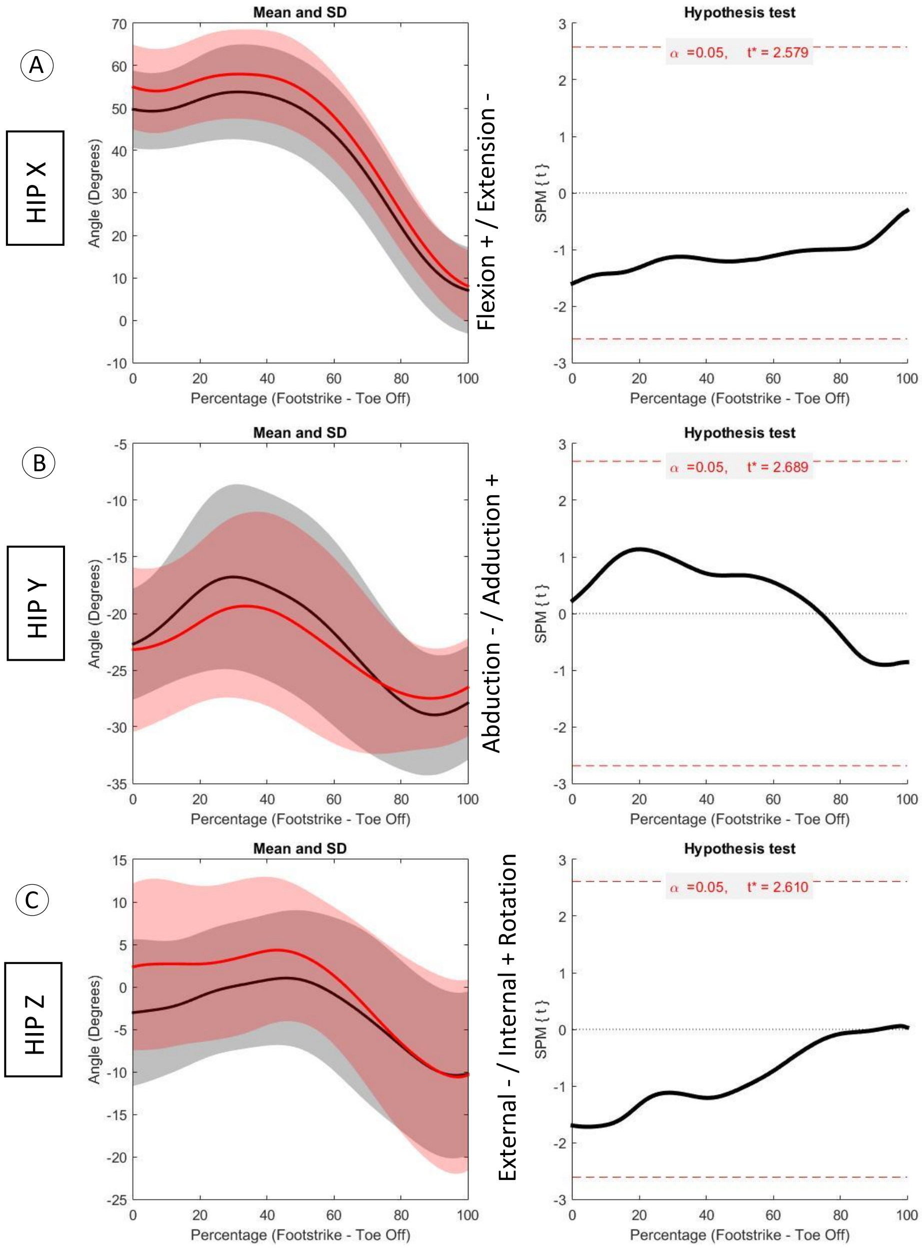


Figure K.35 Cutting- Hip angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

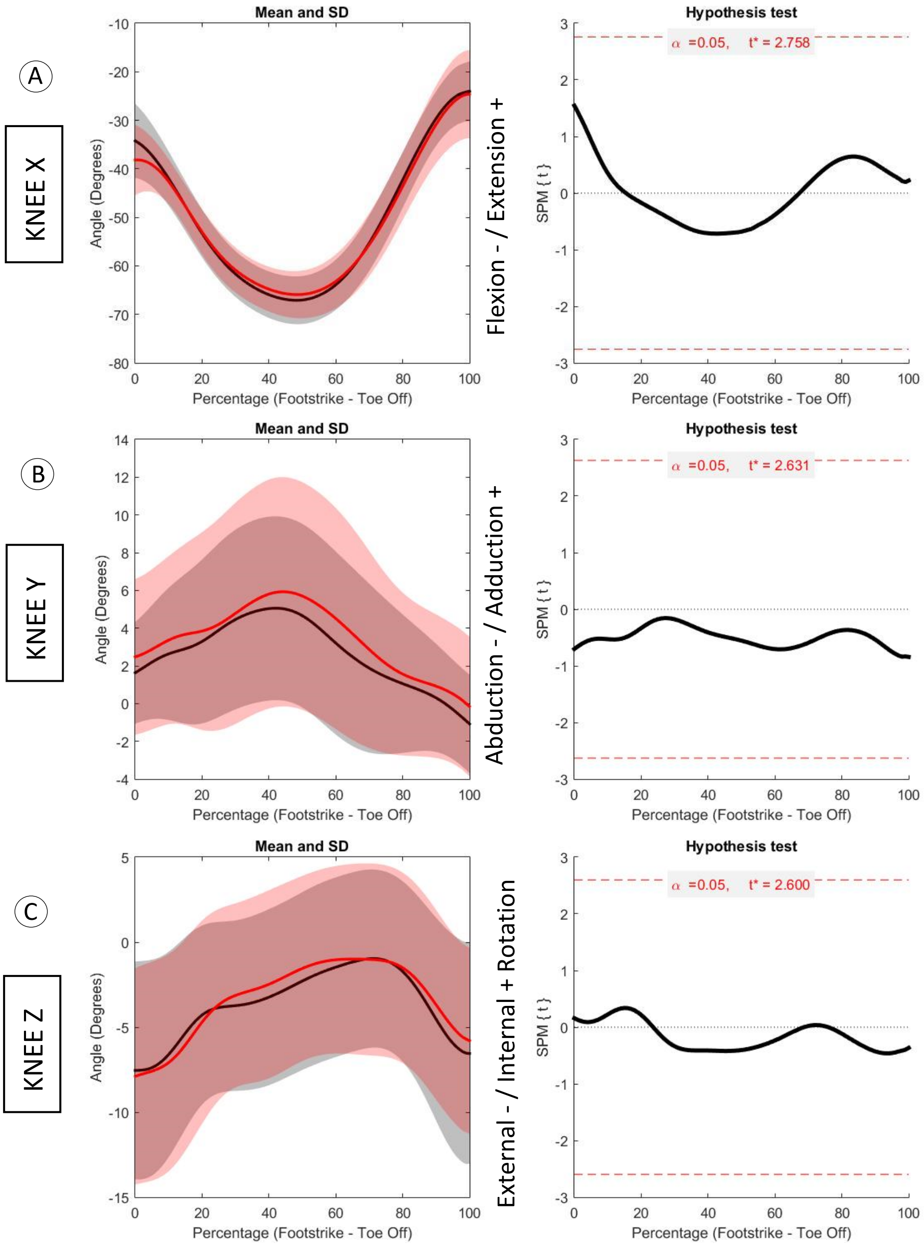


Figure K.36 Cutting- Knee angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

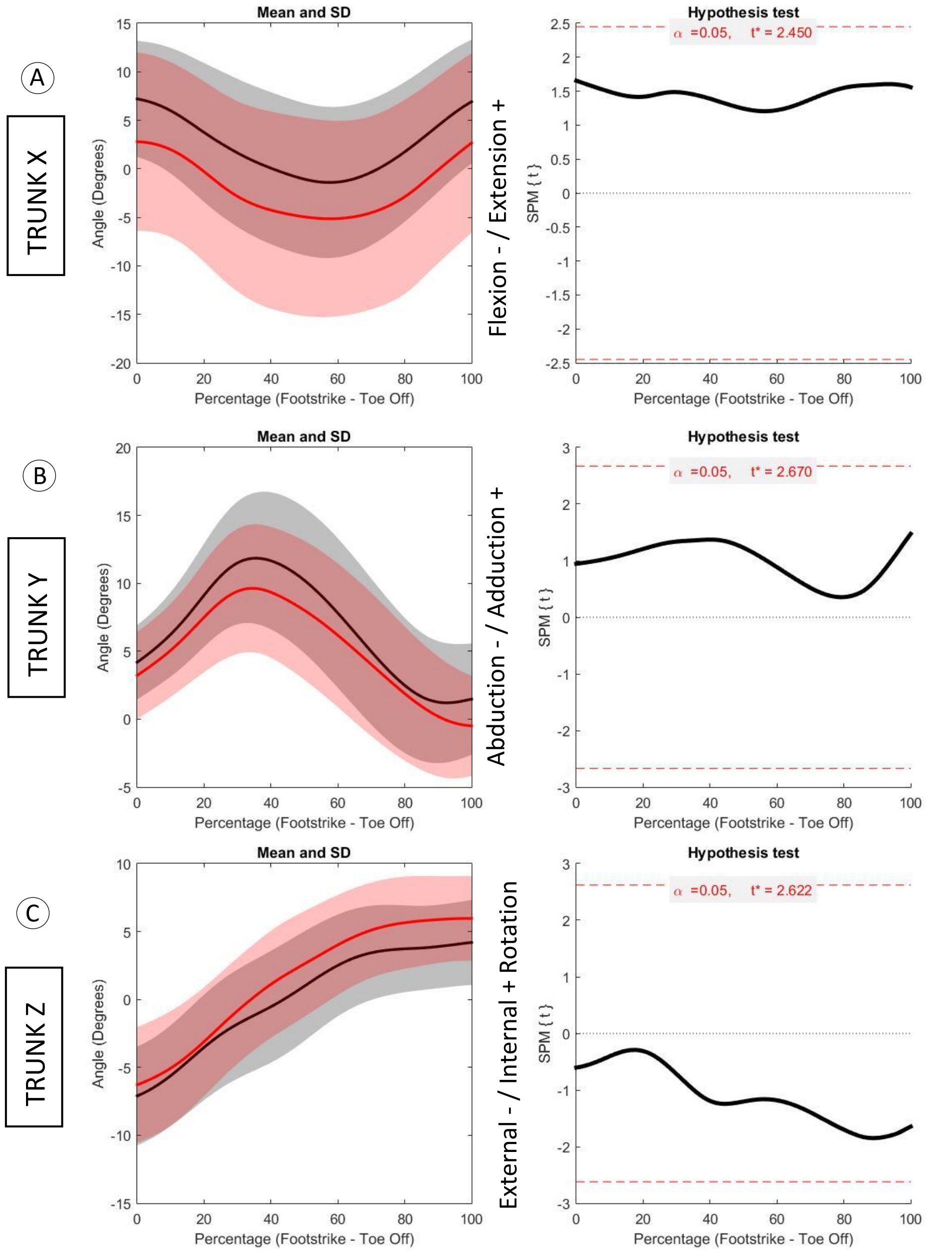


Figure K.37 Cutting- Trunk angles (x, y, z) IC to 200 ms post IC - means and standard deviations (Matched Control - — Unaffected - —) and *t*-test output.

Appendix L Study 3 Part 2 – Cutting Discrete

Angular Displacement

Table L.1 Peak angular displacement from 200 ms pre-initial contact to initial contact (pre-landing phase) (degrees)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	11.00	8.51	9.55	10.63	13.38	11.46	13.73	10.21
	Plantarflexion	0.34	8.71	-2.05	12.31	4.86	13.33	3.11	11.47
FFHFA Frontal	Inversion	12.92	9.61	11.16	9.03	10.79	9.60	8.00	8.14
	Eversion	6.11	7.94	4.59	9.37	4.02	8.96	0.85	6.55
FFHFA Transverse	Adduction	-0.18	6.05	-0.35	7.54	-3.60	6.62	-0.97	6.95
	Abduction	-5.23	7.20	-4.61	7.83	-7.51	6.46	-5.13	6.98
FFTBA Sagittal	Dorsiflexion	4.22	7.30	3.64	9.53	6.50	8.59	5.84	8.91
	Plantarflexion	-18.88	9.79	-21.77	14.39	-13.17	14.25	-15.57	12.92
FFTBA Frontal	Inversion	11.45	5.50	14.65	5.55	13.18	5.17	11.12	6.02
	Eversion	3.30	5.30	4.86	7.00	3.98	4.99	1.58	4.53
FFTBA Transverse	Internal Rotation	12.21*	5.40	10.27	4.58	8.18*	5.17	9.03	7.03
	External Rotation	3.29	5.47	0.32	4.62	0.44	5.73	0.27	7.17
HFTBA Sagittal	Dorsiflexion	-4.72	7.48	-4.80	5.39	-5.06	6.73	-7.06	5.34
	Plantarflexion	-16.54	7.31	-18.06	6.99	-15.90	6.64	-17.53	5.58
HFTBA Frontal	Inversion	-0.72	10.25	3.98	8.38	1.59	8.70	2.98	5.98
	Eversion	-5.27	9.97	-0.87	8.21	-2.68	9.37	-1.30	6.10
HFTBA Transverse	Internal Rotation	13.52	6.81	10.62	6.92	12.02	7.54	10.96	6.88
	External Rotation	6.77	6.30	3.41	6.46	5.65	6.96	3.85	7.07
HIP Sagittal	Extension	64.21	9.73	64.07	12.33	66.21	12.22	67.69	12.60
	Flexion	49.53	7.93	49.35	9.65	51.75	12.09	51.75	12.59
HIP Frontal	Adduction	-13.90	5.19	-14.95	6.19	-11.43	5.71	-11.52	8.31
	Abduction	-23.46	6.12	-23.79	5.08	-21.78	6.09	-23.89	7.79
HIP Transverse	Internal Rotation	1.98	5.50	1.02	7.82	3.29	5.44	4.96	8.33
	External Rotation	-5.88	6.25	-5.91	8.31	-5.13	6.62	-4.09	8.56
KNEE Sagittal	Extension	-34.56	8.95	-32.68	8.59	-35.45	6.78	-37.23	6.59
	Flexion	-85.69	10.42	-83.30	10.82	-85.94	8.10	-87.77	8.03
KNEE Frontal	Adduction	8.22	2.57	8.74	5.13	9.09	6.57	9.24	7.27
	Abduction	0.02	2.61	0.73	3.21	0.98	5.84	1.53	4.88
KNEE Transverse	Internal Rotation	-4.15	5.56	-4.98	5.04	-3.02	5.04	-2.97	3.97
	External Rotation	-11.69	4.66	-12.39	4.97	-11.26	4.30	-11.46	6.60
TRUNK Sagittal	Extension	8.42	5.42	8.87	6.13	3.94	9.18	3.94	8.79
	Flexion	2.27	6.88	2.88	7.49	-2.10	10.36	-2.26	11.05
TRUNK Frontal	Adduction	3.19	2.74	4.26	2.86	3.47	3.67	3.43	3.13
	Abduction	-5.28	5.06	-4.46	3.06	-5.43	4.54	-6.02	3.31
TRUNK Transverse	Internal Rotation	-2.31	3.51	-4.81	3.84	-2.96	4.58	-3.73	4.86
	External Rotation	-6.48	3.44	-8.72	4.19	-7.01	3.68	-8.51	4.47

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table L.2 Angular displacement at initial contact (degrees)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	2.08	8.98	-0.40	12.33	6.09	13.17	4.35	11.65
FFHFA Frontal	12.62	9.58	10.73	9.04	10.48	9.59	7.12	8.30
FFHFA Transverse	-2.38	6.57	-1.88	8.42	-5.23	7.23	-3.16	7.20
FFTBA Sagittal	-16.60	9.74	-20.00	13.61	-11.92	13.61	-13.81	12.32
FFTBA Frontal	10.38	6.21	14.15 †	5.64	12.67	5.31	9.98 †	6.12
FFTBA Transverse	10.59	6.15	9.41	5.11	6.53	6.14	6.82	6.61
HFTBA Sagittal	-15.19	6.95	-17.13	6.86	-15.13	6.32	-16.53	5.13
HFTBA Frontal	-2.56	10.47	3.39	8.66	0.88	8.65	2.05	6.54
HFTBA Transverse	12.75	7.11	10.30	6.98	10.79	7.86	9.60	7.39
HIP Sagittal	50.65	7.67	49.70	9.42	53.60	11.87	54.96	10.28
HIP Frontal	-22.78	6.23	-22.69	5.05	-20.87	6.75	-23.18	7.46
HIP Transverse	-1.60	7.59	-3.16	8.90	1.10	5.48	2.36	10.12
KNEE Sagittal	-35.87	7.93	-34.24	7.86	-36.72	7.32	-38.19	7.48
KNEE Frontal	1.09	2.55	1.65	2.78	2.06	5.00	2.48	4.26
KNEE Transverse	-6.93	5.46	-7.54	6.58	-6.69	5.43	-7.89	6.55
TRUNK Sagittal	6.85	6.27	7.21	6.17	2.25	9.59	2.80	9.48
TRUNK Frontal	3.06	2.68	4.19	2.82	3.26	3.62	3.22	3.30
TRUNK Transverse	-4.36	4.14	-7.10	3.74	-4.85	4.84	-6.28	4.38

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table L.3 Peak angular displacement from initial contact to 200 ms post initial contact (impact phase) (degrees)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	20.97	10.59	18.99	9.82	22.68	9.23	22.98	9.58
	Plantarflexion	2.05	8.98	-0.49	12.26	5.80	12.83	4.07	11.45
FFHFA Frontal	Inversion	16.17	8.62	14.50	7.40	15.01	9.51	11.97	9.00
	Eversion	10.38	7.48	8.50	8.02	9.33	9.13	5.64	8.28
FFHFA Transverse	Adduction	-1.93	6.61	-1.58	8.61	-4.40	7.20	-2.51	6.97
	Abduction	-8.77	7.44	-7.73	9.74	-10.13	7.12	-7.58	6.97
FFTBA Sagittal	Dorsiflexion	20.45	7.38	19.95	7.41	20.04	6.41	21.07	6.69
	Plantarflexion	-16.91	9.45	-20.33	13.20	-12.67	13.02	-14.39	12.03
FFTBA Frontal	Inversion	19.54	4.46	20.21	5.55	20.64	5.72	18.36	7.77
	Eversion	10.22	6.07	13.09	5.05	12.44	5.38	9.54	6.32
FFTBA Transverse	Internal Rotation	11.90	5.53	10.71	5.02	9.13	5.70	8.90	6.63
	External Rotation	4.55	5.53	1.16	5.18	1.95	5.32	1.95	6.03
HFTBA Sagittal	Dorsiflexion	4.38	7.27	3.68	7.05	1.39	7.43	0.85	5.51
	Plantarflexion	-15.54	7.00	-17.28	6.67	-15.61	6.18	-16.85	4.95
HFTBA Frontal	Inversion	1.91	8.94	5.56	8.17	3.31	9.37	5.08	6.38
	Eversion	-3.52	9.57	1.45	7.76	-0.46	9.29	0.84	6.57
HFTBA Transverse	Internal Rotation	16.35	6.87	13.68	6.18	15.67	7.39	13.65	7.42
	External Rotation	11.04	6.49	7.58	6.13	9.41	7.44	8.71	7.29
HIP Sagittal	Extension	55.46	8.75	55.18	11.33	59.09	12.73	59.96	10.38
	Flexion	46.11	9.38	46.06	9.91	48.16	14.04	48.32	11.51
HIP Frontal	Adduction	-17.18	6.41	-16.00	7.94	-15.21	6.93	-17.05	6.76
	Abduction	-23.95	6.17	-23.91	6.16	-23.40	7.76	-26.01	8.04
HIP Transverse	Internal Rotation	3.18	7.52	2.23	8.24	4.76	5.56	6.55	9.36
	External Rotation	-3.66	7.53	-5.13	7.97	-3.46	6.88	-1.73	9.12
KNEE Sagittal	Extension	-35.51	7.62	-33.99	7.55	-35.98	6.24	-37.01	6.52
	Flexion	-65.87	4.61	-66.84	4.72	-66.88	8.59	-65.81	4.82
KNEE Frontal	Adduction	5.00	3.03	5.67	4.70	6.70	5.82	6.40	5.97
	Abduction	0.73	2.56	0.95	3.25	1.34	5.57	1.76	4.63
KNEE Transverse	Internal Rotation	-1.87	5.09	-1.97	5.50	-0.75	4.82	-1.44	5.26
	External Rotation	-7.63	5.50	-8.18	6.31	-7.56	5.26	-8.44	6.35
TRUNK Sagittal	Extension	6.92	6.26	7.45	6.27	2.42	9.52	3.23	9.47
	Flexion	-0.97	8.04	-0.81	8.14	-5.43	10.59	-5.12	10.61
TRUNK Frontal	Adduction	11.66	2.77	12.25	5.04	11.57	4.23	10.35	4.66
	Abduction	2.72	2.57	3.84	2.55	2.95	3.58	2.58	3.37
TRUNK Transverse	Internal Rotation	3.21	5.22	0.68	4.15	2.38	5.54	2.73	4.33
	External Rotation	-4.61	4.08	-7.18	3.75	-5.06	4.98	-6.41	4.33

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Angular Velocity

Table L.4 Peak angular velocity from 200 ms pre-initial contact to initial contact (pre-landing phase) (degrees/second)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	96.19	46.03	99.68	53.38	84.14	39.08	82.82	35.76
	Plantarflexion	-116.63	35.49	-122.18	50.58	-103.77 ‡	42.59	-129.38 ‡	41.33
FFHFA Frontal	Inversion	103.89	20.51	95.06	28.19	97.51	27.58	107.88	28.37
	Eversion	-43.29	17.66	-43.68	26.77	-29.99 ‡	22.98	-42.31 ‡	20.87
FFHFA Transverse	Adduction	57.58	29.13	52.35	25.01	60.91	43.99	59.80	48.76
	Abduction	-64.98	22.97	-55.60	18.35	-50.61	22.97	-55.38	15.71
FFTBA Sagittal	Dorsiflexion	160.77	82.29	167.40	94.19	135.33	80.73	144.58	66.34
	Plantarflexion	-249.04	65.17	-255.63	64.75	-213.17 ‡	74.15	-252.57 ‡	65.89
FFTBA Frontal	Inversion	148.55	50.49	151.76	47.44	140.42	40.38	150.50	49.83
	Eversion	-69.97*	43.58	-53.75	25.51	-39.88*‡	33.29	-58.91 ‡	24.16
FFTBA Transverse	Internal Rotation	95.40	28.90	103.12	32.71	96.10	52.53	111.65	72.78
	External Rotation	-100.72	43.05	-83.88	43.43	-85.81	56.40	-112.04	58.84
HFTBA Sagittal	Dorsiflexion	87.31	44.71	86.71	40.96	70.61	45.65	76.57	42.09
	Plantarflexion	-136.91	33.78	-143.65	31.39	-119.66	33.03	-128.34	31.68
HFTBA Frontal	Inversion	69.72	21.40	82.46	32.30	67.81	24.59	76.52	24.94
	Eversion	-48.26	23.47	-36.80	20.56	-33.96	21.41	-47.17	27.80
HFTBA Transverse	Internal Rotation	78.58	24.90	91.37	24.12	80.73	34.85	96.71	46.15
	External Rotation	-60.47	38.29	-56.27	37.62	-63.79 ‡	43.39	-87.76 ‡	54.46
HIP Sagittal	Extension	51.98	79.80	37.60 †	50.52	68.03	80.81	89.07 †	85.36
	Flexion	-127.28	46.59	-122.12	44.07	-129.31	51.03	-132.50	44.00
HIP Frontal	Adduction	31.73	33.38	41.78	31.67	30.89	32.85	21.30	35.42
	Abduction	-77.48	50.10	-67.79 †	35.46	-97.09	35.75	-103.79 †	44.65
HIP Transverse	Internal Rotation	37.37	33.14	36.38	26.95	40.71	48.58	39.02	40.47
	External Rotation	-102.10	30.35	-90.69	36.60	-96.25	48.54	-108.59	35.76
KNEE Sagittal	Extension	424.72	65.06	413.64	50.02	398.00	44.04	414.41	62.35
	Flexion	-106.18	95.79	-115.82	62.88	-105.85	117.48	-110.78	143.11
KNEE Frontal	Adduction	47.25	28.58	45.91	26.41	44.58	25.79	49.03	31.55
	Abduction	-89.56	54.40	-88.34	42.06	-91.98	59.98	-108.75	69.31
KNEE Transverse	Internal Rotation	106.64	34.88	108.75	31.88	95.88	32.31	96.11	30.33
	External Rotation	-41.95	40.27	-36.27	18.58	-42.49	25.11	-48.13	23.57
TRUNK Sagittal	Extension	51.77	30.21	56.55	25.49	54.31	27.49	55.19	25.91
	Flexion	-34.95	37.13	-29.39	31.45	-32.11	24.72	-27.30	30.28
TRUNK Frontal	Adduction	69.43	23.81	74.59	20.84	72.90	26.14	79.87	24.18
	Abduction	12.62	19.64	16.65	16.36	12.72	27.08	7.46	27.58
TRUNK Transverse	Internal Rotation	33.17	30.35	30.98	24.65	34.86	28.82	32.17	21.81
	External Rotation	-30.58	20.15	-27.22 †	15.01	-33.93 ‡	26.22	-48.57 ‡†	28.48

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table L.5 Angular velocity at initial contact (degrees/second)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	78.41*	58.11	79.49	73.97	34.95*	67.26	45.03	65.09
FFHFA Frontal	69.73	45.03	35.40 †	50.95	72.74	34.76	70.72 †	35.65
FFHFA Transverse	-13.24	29.37	-23.53 †	33.38	-7.65	51.97	-1.25 †	29.21
FFTBA Sagittal	110.93	114.26	134.98	127.30	53.07	128.46	69.16	128.78
FFTBA Frontal	118.52	66.10	68.44 †	76.57	108.31	55.22	119.35 †	47.29
FFTBA Transverse	0.28	40.76	-2.82	55.10	5.02	64.30	-7.14	61.13
HFTBA Sagittal	47.91	59.84	61.74	61.82	32.33	69.63	36.37	70.74
HFTBA Frontal	38.12	35.73	29.74	39.69	36.18	28.04	47.66	30.07
HFTBA Transverse	33.01	31.92	40.56	32.55	32.19	43.88	20.52	44.40
HIP Sagittal	-53.00	55.98	-37.82 †	38.49	-51.24	64.27	-68.81 †	42.72
HIP Frontal	-5.92	46.92	10.39	40.34	1.36	50.75	-19.14	58.21
HIP Transverse	20.70	47.81	25.86	30.18	8.89	56.51	17.47	37.50
KNEE Sagittal	-62.25	112.01	-87.49 †	91.62	-30.89 ‡	129.63	22.56 †‡	116.56
KNEE Frontal	26.48	17.14	18.72	13.21	23.71	18.14	20.90	27.71
KNEE Transverse	-7.10	34.54	-12.54	34.50	7.17	48.22	5.29	44.40
TRUNK Sagittal	-21.05	27.40	-14.45	34.02	-11.66	29.52	-2.07	29.22
TRUNK Frontal	38.39	37.00	38.76	32.38	42.55	34.43	41.03	33.68
TRUNK Transverse	25.63	33.70	26.85	25.81	17.51	26.61	23.54	23.13

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table L.6 Peak angular velocity from initial contact to 200 ms post-initial contact (impact phase) (degrees/second)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	Dorsiflexion	173.78	57.78	187.35	72.83	159.12 ‡	53.59	189.64 ‡	54.76
	Plantarflexion	-8.12	41.19	1.44	31.78	-26.55	38.12	-22.73	44.88
FFHFA Frontal	Inversion	92.43	43.68	77.61	40.61	93.33	37.25	103.27	36.22
	Eversion	-45.10	29.22	-37.05	32.92	-37.90	23.47	-42.22	30.42
FFHFA Transverse	Adduction	37.33	24.60	24.75 †	23.06	51.51	43.35	45.70 †	23.47
	Abduction	-85.69	45.53	-71.38	37.17	-74.31	47.49	-68.40	46.11
FFTBA Sagittal	Dorsiflexion	336.28	108.39	364.60	121.68	291.99	111.82	319.28	92.33
	Plantarflexion	15.03*	54.32	25.40	64.39	-38.18*	91.15	-30.57	100.97
FFTBA Frontal	Inversion	136.66	58.54	99.65	58.68	125.28	58.58	137.58	49.67
	Eversion	0.28	13.70	-24.69	42.58	-11.85	25.67	-12.20	23.70
FFTBA Transverse	Internal Rotation	45.78*	25.60	45.06	35.74	77.00*	35.69	64.27	32.86
	External Rotation	-98.51	37.20	-106.40	49.10	-91.96	49.92	-80.06	36.26
HFTBA Sagittal	Dorsiflexion	188.94	57.79	191.93	67.87	159.79	67.13	164.94	49.53
	Plantarflexion	-7.74	40.41	8.85 †	38.52	-24.83	51.73	-22.63 †	52.23
HFTBA Frontal	Inversion	73.20	31.21	54.35	31.00	59.17	27.73	72.45	26.46
	Eversion	-33.39	25.44	-37.07	23.96	-35.39	20.97	-42.59	38.59
HFTBA Transverse	Internal Rotation	62.72*	23.49	67.93	25.78	95.53*	41.06	76.63	26.62
	External Rotation	-64.87	42.90	-62.18	33.63	-69.31	40.21	-60.08	26.92
HIP Sagittal	Extension	54.73	38.85	70.59	47.49	79.45	48.62	80.19	54.06
	Flexion	-102.55	37.10	-104.06	49.18	-109.21 ‡	73.96	-132.51 ‡	70.79
HIP Frontal	Adduction	60.36	32.64	69.62	43.75	74.94	49.73	76.85	40.08
	Abduction	-66.16	38.14	-70.03	31.19	-64.96	29.46	-72.59	42.78
HIP Transverse	Internal Rotation	86.90	30.93	99.58	64.67	87.69	39.74	91.77	40.45
	External Rotation	-57.63*	31.19	-61.58	53.65	-80.91*	37.02	-83.11	48.53
KNEE Sagittal	Extension	43.98	61.21	36.50	68.58	62.69 ‡	114.17	98.17 ‡	124.63
	Flexion	-262.98	59.94	-275.95	49.75	-279.62	61.07	-267.59	42.33
KNEE Frontal	Adduction	69.78	29.16	69.66	30.47	75.43	26.78	78.62	41.38
	Abduction	-11.23	15.19	-10.90	16.00	-15.17	27.04	-16.83	22.28
KNEE Transverse	Internal Rotation	65.35	29.55	77.42	40.53	64.56	32.91	67.23	27.60
	External Rotation	-56.53	26.25	-64.20	40.40	-63.63	46.78	-49.31	17.69
TRUNK Sagittal	Extension	8.76	14.82	14.67	16.14	14.01	18.59	22.46	18.87
	Flexion	-64.73	26.47	-68.96	33.52	-72.18	29.15	-76.50	31.08
TRUNK Frontal	Adduction	87.77	28.94	78.72	42.69	87.65	24.23	79.50	29.27
	Abduction	-50.49	25.58	-41.17	22.82	-36.06	25.30	-44.30	21.82
TRUNK Transverse	Internal Rotation	71.24	25.25	64.79	20.31	68.44	15.68	80.64	31.58
	External Rotation	5.64	19.01	5.82	16.43	-5.06	16.54	2.08	19.50

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Angular Acceleration

Table L.7 Peak angular acceleration from 200 ms pre-initial contact to initial contact (pre-landing) (degrees /second²)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	+ VE	4906.81	1236.44	5000.78	1796.80	4083.47	1713.14	5109.06	1483.39
	- VE	-2315.90	641.14	-2352.56	1107.53	-2440.23	1418.07	-2744.47	1224.41
FFHFA Frontal	+ VE	2430.49	591.29	2140.00	718.49	2448.82	817.04	2539.92	717.34
	- VE	-1882.17	865.17	-2052.91	1063.41	-1822.01	613.64	-2248.86	1076.01
FFHFA Transverse	+ VE	1898.21	535.14	1675.90	667.50	1904.51	1050.80	2076.97	1194.04
	- VE	-2658.52	813.60	-2120.52	676.09	-2509.56	1262.23	-2377.20	1082.32
FFTBA Sagittal	+ VE	8839.02	2451.11	9566.68	2758.22	7443.29 ‡	3131.21	9195.40 ‡	2184.69
	- VE	-4164.26	902.98	-3972.32	1397.44	-3740.84 ‡	2408.71	-4400.89 ‡	2531.33
FFTBA Frontal	+ VE	3324.87	908.14	2871.66	761.46	3002.30	912.72	3209.71	1061.12
	- VE	-1897.64	614.21	-2523.31	1398.59	-2045.46	901.84	-2201.88	1218.68
FFTBA Transverse	+ VE	3256.12	1271.97	3617.66	2656.23	3212.57	1634.08	3520.83	2032.28
	- VE	-3487.59	1081.17	-3893.25	1227.85	-3649.17	2302.72	-4388.86	2847.03
HFTBA Sagittal	+ VE	4534.39	1539.54	4997.91	1691.94	4132.17	1887.90	4589.39	1255.92
	- VE	-2518.15	708.18	-2436.22	693.28	-2262.04	856.00	-2498.67	1008.98
HFTBA Frontal	+ VE	2220.78	602.18	1899.36	572.96	1889.04	962.00	2220.78	972.88
	- VE	-1724.60	670.36	-1851.23	1057.11	-1831.75	916.73	-1859.82	992.36
HFTBA Transverse	+ VE	2661.39	1204.93	2746.19	1074.30	3076.78	1443.85	3281.90	1435.83
	- VE	-2219.07	814.17	-2435.07	965.43	-2797.18	1347.02	-3495.62	2279.80
HIP Sagittal	+ VE	1886.75	687.55	1928.00	779.80	2103.33	889.80	2062.65	1067.99
	- VE	-2165.21	777.50	-2169.79	898.40	-2421.89	1000.38	-2690.04	1176.86
HIP Frontal	+ VE	1629.49 *	689.84	1739.50 †	450.34	2261.46 *	1093.78	2223.65 †	718.49
	- VE	-1513.75	951.11	-1307.49 †	692.13	-1579.07	758.60	-1894.20 †	823.34
HIP Transverse	+ VE	2110.78	747.71	2156.61	1054.82	2019.10	845.69	2561.12	1144.77
	- VE	-1310.93	675.52	-1261.65	717.34	-1411.77	997.52	-1447.29	834.80
KNEE Sagittal	+ VE	4108.11	2028.27	3858.30	1873.00	4053.68 ‡	1915.97	4607.73 ‡	2171.51
	- VE	-7369.96	2072.96	-7289.17	1605.43	-7057.69	1604.28	-7118.43	1838.05
KNEE Frontal	+ VE	2362.88	1383.69	2532.47	976.32	2413.30	905.85	2869.95	1606.57
	- VE	-1566.47	1684.50	-1541.26	984.91	-1697.10	1156.80	-1917.69	1528.65
KNEE Transverse	+ VE	2547.94	1237.02	2534.77	955.12	2572.58	1057.68	2739.88	976.89
	- VE	-2759.94	1275.98	-2852.76	1266.81	-2302.72	1033.04	-2589.77	911.58
TRUNK Sagittal	+ VE	714.48	336.33	633.12 †	200.54	850.27	421.70	1121.28 †	523.68
	- VE	-1135.60	559.21	-1199.77	461.80	-1208.94	473.26	-1136.75	348.93
TRUNK Frontal	+ VE	759.17	461.80	708.75	374.71	777.50	291.06	877.20	333.46
	- VE	-739.69	337.47	-807.30	322.58	-734.53	394.77	-887.51	540.87
TRUNK Transverse	+ VE	862.87	387.32	843.97 †	245.23	906.42 ‡	563.79	1286.86 ‡†	732.24
	- VE	-331.74	211.42	-328.30 †	146.68	-497.90	615.36	-639.99 †	475.55

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table L.8 Angular acceleration at initial contact (degrees /second^2)

	Control group				CAI group			
	Matched Affected		Matched Unaffected		Affected		Unaffected	
	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	3901.27	1493.70	4367.66	2036.86	3237.78 ‡	1462.76	4103.52 ‡	1841.49
FFHFA Frontal	-652.03 *	1125.86	-557.49	1330.41	309.97 *	960.85	-2.86	1664.44
FFHFA Transverse	-1931.44	1219.83	-1362.49	921.32	-1290.30	1274.83	-1200.92	1267.38
FFTBA Sagittal	8050.06	2739.31	8784.59	3285.34	6435.46	3138.66	7669.61	2440.23
FFTBA Frontal	51.57	1518.91	-984.91	1901.07	442.32	1879.87	110.58	2047.18
FFTBA Transverse	-2066.09 *	1500.00	-2190.99 †	1533.81	-709.89 *	1785.91	-1182.58 †	911.58
HFTBA Sagittal	3778.66	1872.43	4420.37	1940.04	3314.56	2159.48	3629.69	1415.21
HFTBA Frontal	258.40	1330.41	-580.98	1278.84	-93.39	1223.84	26.93	1192.33
HFTBA Transverse	-293.35 *	1484.53	-436.59 †	1147.63	813.03 *	1556.73	358.10 †	900.12
HIP Sagittal	1168.83	1149.35	1211.23	1050.23	1667.88	1025.59	1524.64	1179.72
HIP Frontal	597.02	1031.90	536.29	938.50	1067.42	1191.18	898.97	1225.56
HIP Transverse	632.55	1195.19	948.25	1285.72	263.56	884.07	277.88	1086.90
KNEE Sagittal	-5817.24	2208.75	-5381.22	1350.46	-5950.17	1763.56	-6399.94	2172.08
KNEE Frontal	233.77	1200.92	400.50	1022.16	508.21	1344.73	460.09	1390.57
KNEE Transverse	-672.08	2250.01	-190.22	1296.03	-709.89	1645.53	-1055.39	1371.09
TRUNK Sagittal	-584.42	633.69	-450.92	630.83	-551.19	585.56	-568.95	665.78
TRUNK Frontal	185.07	359.82	108.86	468.68	167.88	373.57	53.86	525.40
TRUNK Transverse	285.33	244.65	443.47	249.24	323.15	334.61	277.88	535.72

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Table L.9 Peak angular acceleration from initial contact to 200 ms post-initial contact (impact phase) (degrees /second^2)

		Control group				CAI group			
		Matched Affected		Matched Unaffected		Affected		Unaffected	
		Mean	St.Dev	Mean	St.Dev	Mean	St.Dev	Mean	St.Dev
FFHFA Sagittal	+ VE	4230.15	1321.81	4728.05	1542.98	4117.85	1320.67	4905.66	1278.84
	- VE	-2727.85	1006.69	-2561.69	1101.80	-2499.24 ‡	980.90	-2918.65 ‡	749.43
FFHFA Frontal	+ VE	1893.63	1047.94	1705.12	909.86	1866.12	886.94	2034.57	631.40
	- VE	-2993.13	1035.91	-2445.38	1078.31	-2857.34	1036.48	-3072.77	891.52
FFHFA Transverse	+ VE	2106.19	643.43	1730.91	593.58	2045.46	1074.30	1866.70	803.86
	- VE	-2543.93	1051.38	-1940.61	703.02	-2700.92	1929.72	-2127.39	1019.86
FFTBA Sagittal	+ VE	8654.53	2434.50	9429.74	2338.81	7784.20	2421.32	8632.18	1753.25
	- VE	-3898.98	1713.72	-4265.10	1814.56	-3525.98 ‡	1869.56	-4114.41 ‡	1298.32
FFTBA Frontal	+ VE	1381.97	949.39	1571.05	758.02	1700.54	669.79	1636.94	957.41
	- VE	-2832.13	983.20	-2899.74	824.49	-2944.43	1132.16	-3264.71	690.99
FFTBA Transverse	+ VE	2881.98	882.93	2838.43	1069.71	2934.12	1440.42	2652.79	1137.89
	- VE	-3213.72	865.74	-3362.12 †	1182.58	-3381.02 ‡	1592.25	-2673.42 ‡†	658.90
HFTBA Sagittal	+ VE	4650.13	1302.91	4889.05	1466.20	4349.90	1541.26	4419.80	1163.68
	- VE	-2886.56	1204.93	-2490.07	1092.06	-2615.55	1065.13	-2743.32	990.64
HFTBA Frontal	+ VE	2039.73	1130.45	1602.56	800.99	1746.95	814.75	1542.40	536.29
	- VE	-1993.32	741.98	-1949.20	725.36	-1973.27	854.85	-2165.21	631.97
HFTBA Transverse	+ VE	2368.03	1040.49	2370.90	1009.55	3073.92	1773.88	2641.34	1183.73
	- VE	-2989.69	1188.31	-2832.13	958.56	-3350.66	1677.62	-2782.28	959.13
HIP Sagittal	+ VE	2290.69	642.86	2301.57 †	754.01	2846.45	1129.87	3084.80 †	1336.71
	- VE	-2145.73	491.60	-2408.14	1247.33	-2596.64 ‡	877.77	-3064.75 ‡	1015.85
HIP Frontal	+ VE	2034.57	962.57	2005.35 †	817.61	2364.60	778.65	2703.79 †	865.74
	- VE	-2287.82	840.53	-2364.60	1182.01	-2470.59	895.53	-2278.08	776.36
HIP Transverse	+ VE	1985.87	927.62	2006.50	1027.31	2205.31	848.55	2388.66	1184.30
	- VE	-2346.84	741.98	-2568.00	1320.67	-2714.10	967.15	-2983.39	1091.48
KNEE Sagittal	+ VE	3167.31	692.71	3201.69	748.86	3646.30	896.11	3653.75	1009.55
	- VE	-5924.38	2219.07	-5437.37 †	1475.37	-6421.71	2028.27	-7106.97 †	2771.97
KNEE Frontal	+ VE	1631.78	904.70	1601.99	698.44	1791.07	888.66	1827.16	1099.51
	- VE	-1885.03	898.97	-1862.11	787.24	-1862.11	1087.47	-1966.96	1212.38
KNEE Transverse	+ VE	2826.97	1391.71	2730.72	1042.78	2706.08	1231.29	2678.58	964.29
	- VE	-2684.88	1228.42	-2786.87	1050.23	-2836.71	1297.75	-2793.74	948.82
TRUNK Sagittal	+ VE	1123.00	391.90	1224.41	696.72	1377.96	550.04	1593.97	645.72
	- VE	-1065.70	450.92	-1343.59	965.43	-1339.58	501.91	-1595.11	638.27
TRUNK Frontal	+ VE	1054.24	474.98	968.87	501.34	955.12	542.59	956.27	504.20
	- VE	-1886.75	582.70	-1652.98	589.00	-1640.38	308.82	-1886.18	454.36
TRUNK Transverse	+ VE	817.61	269.86	835.37 †	281.90	893.81	296.79	1186.02 †	636.56
	- VE	-859.44	383.31	-853.71	389.04	-925.33	395.34	-1182.01	975.17

*Notes: * = significant difference between CAI group affected and control group matched affected. † = significant difference between CAI group unaffected and control group matched unaffected. ‡ = significant difference between CAI unaffected and CAI affected.*

Appendix M Study 3 Part 3 – Cutting Regression

SPSS correlation outputs

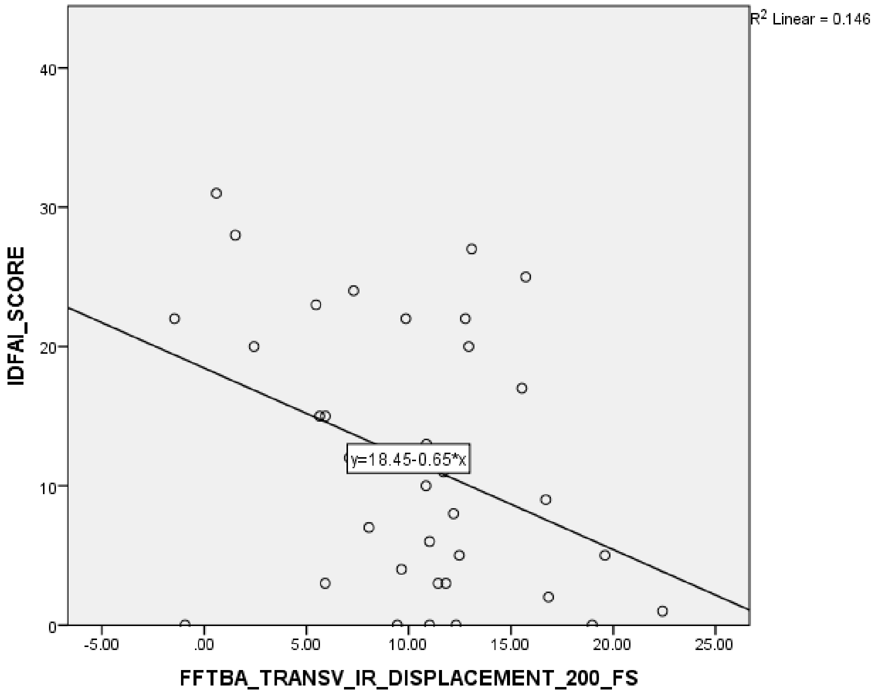


Figure M.1 Pearson's correlation FFTBA transverse internal rotation displacement 200 ms pre IC to IC and IdFAI score

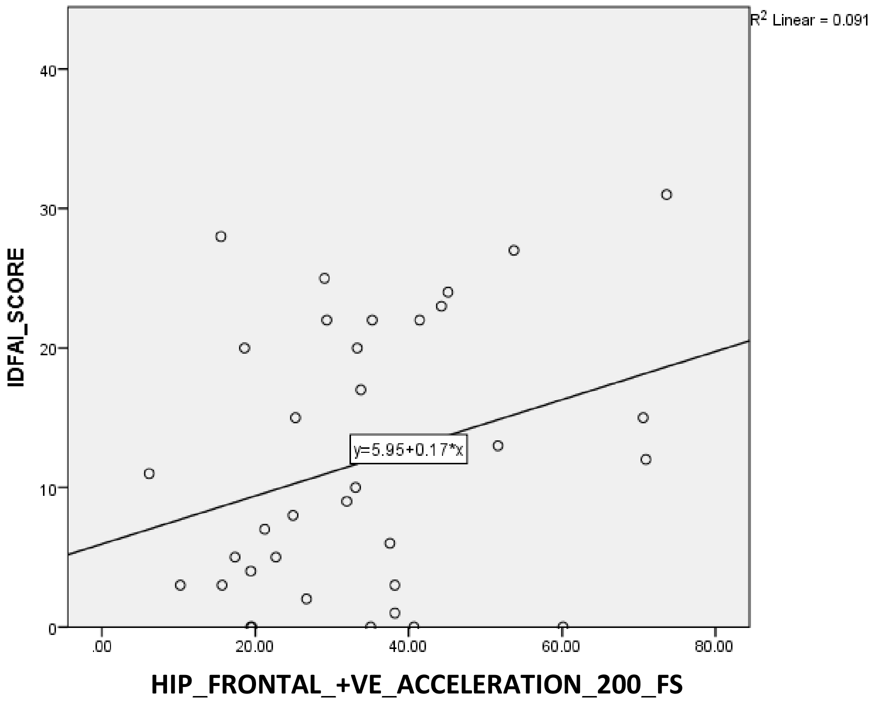


Figure M.2 Pearson's correlation hip frontal peak positive acceleration 200 ms pre IC to IC and IdFAI score

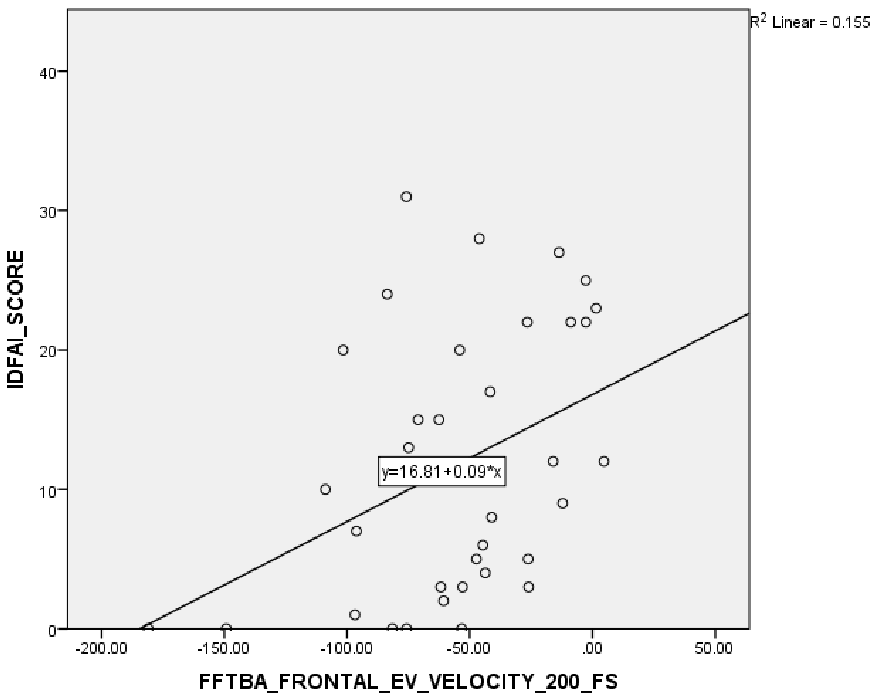


Figure M.3 Pearson's correlation FFTBA frontal eversion velocity 200 ms pre IC to IC and IdFAI score

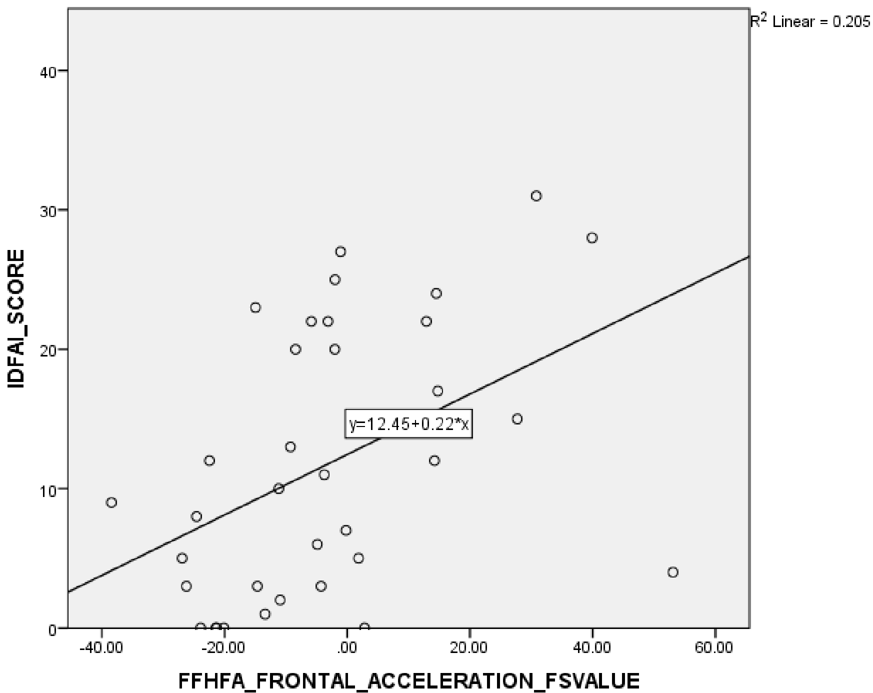


Figure M.4 Pearson's correlation FFHFA frontal acceleration at IC and IdFAI score

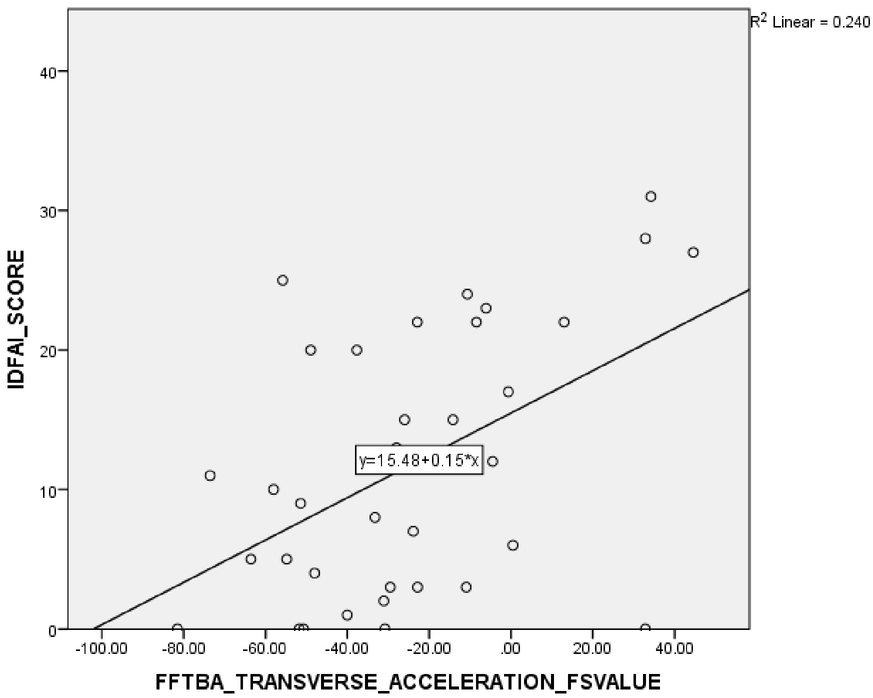


Figure M.5 Pearson's correlation FFTBA transverse acceleration at IC and IdFAI score

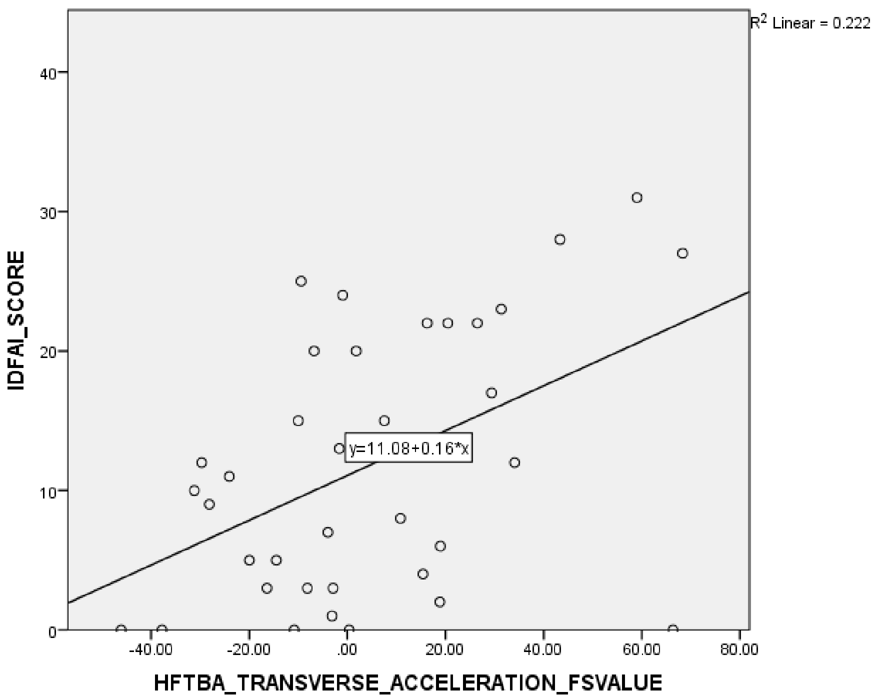


Figure M.6 Pearson's correlation HFTBA transverse acceleration at IC and IdFAI score

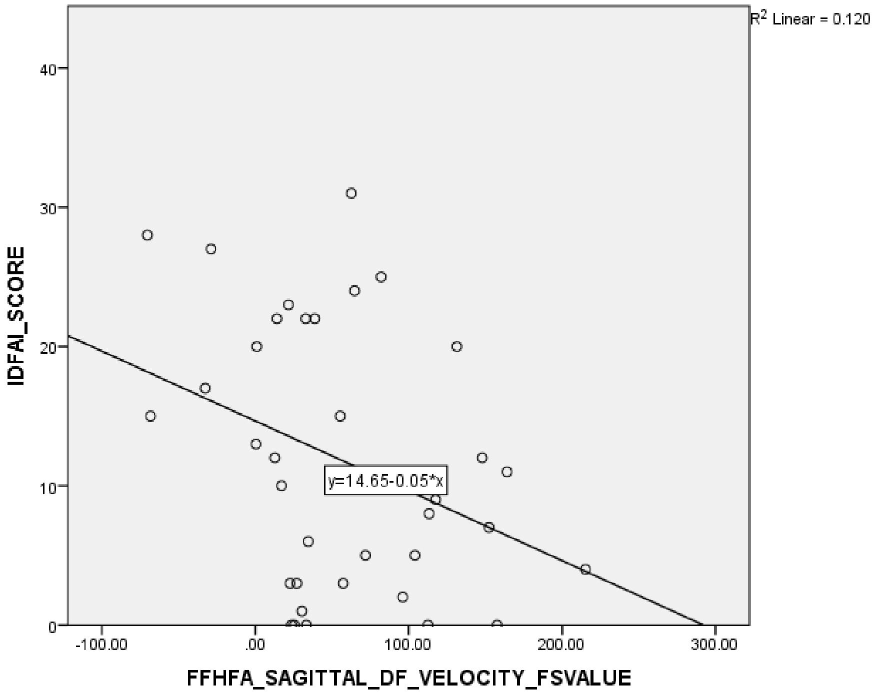


Figure M.7 Pearson's correlation FFHFA sagittal dorsiflexion velocity at IC and IdFAI score

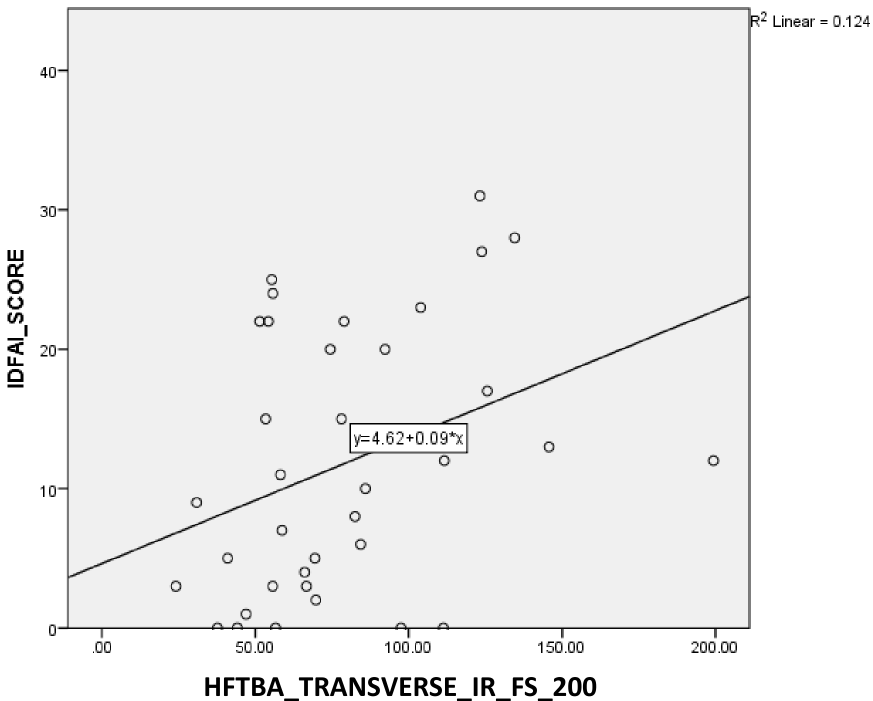


Figure M.8 Pearson's correlation HFTBA transverse internal rotation velocity footstrike to 200 ms post footstrike

Table M.1 R values for regression analysis model with IdFAI as dependent variable

	R	R Square	Adj R Square	F	Sig
FFTBA transverse acceleration ICVALUE	0.490	0.240	0.218	10.742	0.002
FFTBA transverse acceleration ICVALUE, FFTBA frontal Eversion velocity 200-IC	0.600	0.360	0.321	6.191	0.018
FFTBA transverse acceleration ICVALUE, FFTBA frontal Eversion velocity 200-IC, FFHFA frontal acceleration ICVALUE	0.666	0.443	0.391	4.756	0.037