The effect of muscular fatigue on weight distribution, center of pressure position, and lower-limb muscle activation during static squats in healthy individuals

Ioannis Makris, BSc, PT

School of Physical and Occupational Therapy
Faculty of Medicine, McGill University, Montreal, Quebec, Canada
August 2012

A thesis submitted to the Faculty of Graduate Studies and Research in partial fulfillment of the requirement of the degree of MSc in Rehabilitation Science

© Ioannis Makris, 2012

Table of Contents

List of figures

List of tables

Abbreviations

ACL: anterior cruciate ligament

ANOVA: analysis of variance

BF: biceps femoris

COM: center of mass

COP: center of pressure

EMG: electromyography

GL: gastrocnemius lateralis

GM: gastrocnemius medialis

JPS: joint position sense

RF: rectus femoris

ST: semitendinosus

TA: tibialis anterior

TTDPM: threshold to detect passive motion

VM: vastus medialis

Acknowledgements

I would like to deeply thank my supervisor, Dr. Patricia McKinley, for her guidance and comments on every aspect of my work and also for giving me the opportunity to meet with Dr. St-Onge. Above all, I truly appreciate the fact that she took the time to correct my grammatical and syntactical errors that are unavoidable when assembling a document like this in a second language.

I would also like to express my sincere gratitude to my co-supervisor, Dr. Nancy St-Onge, for her continuous guidance and support throughout the last three years. I would like to thank her because by her I discovered and loved the demanding area of research. I would also like to thank her for providing me with a solid and interesting initial research protocol and for supplying me the lab facilities and equipment which were basic tools for the completion of my thesis.

I would also like to thank the other members of my supervisory committee, Dr. Barbara Mazer for her comments and her valuable help on statistics and Dr. Julie Côté for providing interesting comments on the methodology of this project.

I would like also to thank some people from Concordia University which was my base for performing the experiments and writing this thesis. In particular, Dr. Richard DeMont, professor in the Department of Exercise Science, for filling me in on any clinical question that I had on the project. Moreover, I wish to thank two undergraduate students in the Department of Exercise Science, Shennel for helping me gather pilot data for my project and Eva who helped me with the recruitment of the participants and the data collection and analysis.

I would also like to thank the School of Physical and Occupational Therapy for the financial support they provided me by granting me the Principal's Graduate Fellowship.

I am also grateful for having by my side my partner and friend, Maria, who was always there to support and encourage me, especially during times when the burden of work seemed unbearable.

Last but for sure not least, I would like to express my deepest gratitude to my parents who believed in me all these years and always made me challenge myself and push my limits further. With their emotional and financial support, they created a safe environment, even being thousand miles away, which gave me the peace of mind in order to complete this achievement.

Contribution of authors

The basic concepts of this project are stated and interpreted in a manuscript, included in this thesis. A similar version of the manuscript will be submitted in the peer-reviewed *Journal of Orthopaedic and Sports Physical Therapy* and will have as co-authors Mr. Ioannis Makris, Dr. Patricia McKinley and Dr. Nancy St-Onge. In accordance with the guidelines of the Faculty of Graduate and Postdoctoral Studies of McGill University, I would like to declare the contribution of all co-authors. The experimental protocol was initially designed by Dr. St-Onge and was modified and piloted by Ioannis Makris and Dr. St-Onge. Subject recruitment, data collection and data analyses were performed by Mr. Makris under the guidance of Dr. St-Onge. The manuscript was prepared by Mr. Makris with feedback from Dr. St-Onge and Dr. McKinley.

Abstract

Background: Muscular fatigue is a natural consequence of prolonged exercise and has been shown to decrease force production. Muscle fatigue has also been proven to increase muscle reflex time and amplitude and to decrease proprioception, probably as a consequence of affecting several neural receptors. Furthermore, studies on fatigue have concluded that it affects unilateral and bilateral stance. Although balance is decreased during quiet standing after fatigue, we do not know how it is affected in other positions, such as during squats.

Objectives and hypotheses: The objectives were to define and compare, during static squats, the weight distribution, the center of pressure (COP) position and the activity of the muscles in the lower extremities of a healthy population before and after a fatigue protocol. Our first hypothesis was that significant differences in weight distribution and COP position, after fatigue, would be observed. Specifically, we expected weight distribution in the sagittal plane (anteroposterior body axis) and the COP position to be more forward, but expected no difference in the frontal plane (mediolateral body axis). Our second hypothesis was that the activity of the lower limb muscles, after fatigue, would be different. For example we expected the quadriceps and the gastrocnemius to be more active.

Methods: Fifteen healthy, physically active individuals (age 18-30 years) were selected to participate in our study. Data were collected before and after a lower-limb fatigue protocol, while the participants were in different static squatting positions. Pressure distribution was collected using a MatScan and was used to compute weight distribution in two planes (frontal and sagittal) and COP position in the sagittal plane. Electromyographic (EMG) data from 14 lower-limb muscles (seven from each leg) were collected.

Results: The fatigue protocol had a significant effect on the COP position in the sagittal plane, and on the activity of some muscles. However, it did not have an effect on weight distribution whether in the sagittal plane or in the frontal plane. Squat position had a significant effect on the activity of some muscles, but not on weight distribution or on the COP position.

Significance: There are many studies on the effect of fatigue on standing balance and proprioception. Their results show that fatigue affects proprioception and balance but its effect on balance has not been studied during squats, when the knee extensors are more active. We believe that by evaluating weight distribution, COP position and muscle activation during a fatigued state, we may be able to understand the effect of fatigue on balance in positions that are more challenging in healthy individuals and eventually in injured populations. Fatigued conditions, like the protocol we used, can resemble real life conditions during games, where the fatigue state of the participants is usually increased.

Résumé

Contexte : La fatigue musculaire est une conséquence normale de l'exercice prolongé et il a été démontré qu'elle mène à une diminution de la force pouvant être produite. La fatigue musculaire augmente aussi le temps de latence et l'amplitude des réflexes musculaires; elle diminue la proprioception. Ceci découle probablement du fait que la fatigue affecte les récepteurs neuraux. En outre, les études sur la fatigue ont conclu qu'elle affecte l'équilibre en position debout, que ce soit en situation unilatérale ou bilatérale. Bien que la fatigue diminue l'équilibre en position debout, nous ne savons pas si l'équilibre est affecté pour d'autres positions, comme, par exemple, pendant les squats.

Objectifs et hypothèses : Les objectifs de cette thèse consistaient à définir et comparer, pendant des squats statiques, la distribution du poids, la position du centre de pression et le recrutement musculaire des membres inférieurs d'une population saine avant et après un protocole de fatigue musculaire. Notre première hypothèse était que la fatigue musculaire mènerait à des différences significatives de la distribution du poids et de la position du centre de pression. Plus particulièrement, nous croyions que les participants déplaceraient leur poids et leur centre de pression vers l'avant, mais nous ne prévoyions pas que la fatigue affecterait la distribution du poids dans le plan frontal. Notre deuxième hypothèse était que le recrutement musculaire du membre inférieur seraient différents suite à la procédure de fatigue. Par exemple nous nous attendions à ce que les quadriceps et les gastrocnemius soient plus activés.

Méthodologie : Quinze individus sains et physiquement actifs âgés entre 18 et 30 ans ont participé à notre étude. Nous avons enregistré des données avant et après un protocole de fatigue musculaire visant les membres inférieurs, alors que les participants effectuaient des squats statiques dans différentes positions. Nous avons enregistré la distribution de pression à l'aide

d’un MatScan et nous l’avons employée pour calculer la distribution de poids dans le plan frontal et le plan sagittal ainsi que la position du centre de pression dans le plan sagittal. Nous avons aussi enregistré l’activité électromyographique de 14 muscles des membres inférieurs (sept de chaque côté).

Résultats : Le protocole de fatigue a produit un effet significatif sur la position du centre de pression dans le plan sagittal ainsi que sur l'activité de certains muscles. Cependant, il n'a pas affecté la distribution du poids que ce soit dans le plan sagittal ou dans le plan frontal. La position des squats a mené à une modification significative de l'activité de certains muscles, mais n’a démontré aucun effet sur la distribution du poids ou sur la position du centre de pression.

Importance de l’étude : Il y a plusieurs études portant sur l'effet de la fatigue musculaire sur la proprioception ainsi que sur l'équilibre en position debout. Les résultats de ces études démontrent que la fatigue affecte la proprioception et l'équilibre, mais l’effet de la fatigue n'a pas été étudié pendant les squats, alors que les extenseurs du genou sont plus activés. Nous croyons qu'en évaluant la distribution du poids, la position du centre de pression et le recrutement musculaire en situation de fatigue musculaire, nous pourrons mieux comprendre l'effet de la fatigue lors de positions qui sont plus exigeantes, dans un premier temps chez les individus sains et par la suite chez les populations blessées. Le protocole utilisé a permis d’augmenter la fatigue, ce qui s’apparente à des situations de tous les jours ou lors de la pratique sportive alors que la fatigue des participants est souvent augmentée.

Preface

There were many steps involved in the development of this manuscript-based thesis. I wrote the thesis and the manuscript under the guidance of Dr. Nancy St-Onge and Dr. Patricia McKinley. Specific contributions are detailed above in the "Contribution of authors" section. The purpose of this thesis was to investigate the effect of muscle fatigue on COP position, weight distribution and lower-limb muscle activation during squats in healthy individuals. This objective is addressed in the manuscript, a similar version of which will be submitted to the scientific *Journal of Orthopaedic and Sports Physical Therapy.* Additional chapters have been incorporated in this thesis in order to comply with the guidelines of the Faculty of Graduate and Postdoctoral Studies of McGill University.

Organization of thesis

Chapter 1 is an introduction and a literature review on the relationship between muscular fatigue and injuries, and on the effect of muscular fatigue on proprioception and balance.

Chapter 2 contains the rationale, the objectives and the hypotheses of the thesis.

Chapter 3 consists of the manuscript entitled ''Muscle fatigue affects center of pressure position and lower-limb muscle activation during static squats in healthy individuals''. The format of the manuscript follows the style for the *Journal of Orthopaedic and Sports Physical Therapy* to which a similar version will be submitted.

Chapter 4 includes the conclusions of the thesis.

Appendices contain extra information for a better understanding of the study. Appendix A shows antero-posterior distribution graphs for each participant. These graphs were used to decide where to split the data into anterior and posterior parts. Appendix B displays normality graphs with Shapiro-Wilk normality test values; appendix C contains *post hoc* analyses for significant main effects and interactions. In appendix D, tables are shown containing information on how data were distributed across participants. Finally the Matlab script used to convert antero-posterior weight distribution vectors is shown in appendix E.

Chapter 1 – Background

1.1 Introduction

Quoting from a recent review article we read that "the term muscle fatigue is used to denote a transient decrease in the capacity of the muscle to perform physical actions" (Enoka & Duchateau, 2008). According to the authors, during fatigue, muscles decrease their ability to produce maximal force or power. Fatigue can occur anywhere along the pathway involved in muscular contraction and is present due to changes in cortical input, excitatory drive to the lower motor neuron, loss of recruitment of high threshold motor units, loss of positive feedback from muscle spindles type I sensory afferents, changes in motor neuron excitability, changes on the transmission at the neuromuscular junction, impaired interaction between myosin and actin during cross-bridge cycling, impaired reuptake and changes to the contractile apparatus and the metabolic energy supply (Bigland-Ritchie, 1981; Dobkin, 2008; Kent-Braun, 1999; J. L. Taylor, Todd, & Gandevia, 2006). Fatigue can be either central or peripheral depending on which point of the pathway is affected. Both types of fatigue lead to decreased force production but have different mechanisms. Central fatigue arises from changes that happen before the neuromuscular junction and peripheral fatigue from changes that happen after the neuromuscular junction. The following literature review will explore, firstly, the studies that show the relationship between fatigue and increased risk of injuries in the musculoskeletal system. Moreover, it will show how fatigue affects proprioception and balance in standing.

1.2 Muscle fatigue and injury

The hypothesis that there is a connection between fatigue and risk of injuries was examined during soccer matches. The following studies investigated the relationship between the possible level of fatigue at specific time periods of the match and injuries.

Focussing on the relationship between fatigue and the possibility of injury, Carling et al. (Carling, Gall, & Reilly, 2010) scanned through an injury database of an elite soccer association, and concluded that the physical demands of modern soccer are high and players are subjected to fatigue and risk of injury The researchers split each game into six 15-minutes periods and included only injuries after which the player had to leave the field. In order to examine the situation of the player at the moment of the injury, they investigated the characteristics of the running action in the minutes preceding each injury, such as running duration and covered distance. They concluded that players, at the time of the injury, may have experienced transient fatigue due to incomplete recovery between high-intensity bouts, which increased their susceptibility to injury. According to the authors, this incomplete recovery could have affected performance in areas such as proprioceptive ability, dynamic joint stability, force production, neuromuscular responses or running kinematics.

Furthermore, another study on the injury risk during competitive soccer showed that risk is higher during the first and the last 15 minutes of a game (Rahnama, Reilly, & Lees, 2002). The researchers, after watching and analysing 10 soccer matches on TV, likewise split each game into six periods of 15 minutes and recorded the number of injuries in each period. The higher injury risk during the first minutes of a match was attributed to the intense engagements that the players show at the opening period while the higher injury risk during the last 15 minutes of a match was attributed to the possible effect that fatigue has in the closing periods.

The previously mentioned studies have suggested that fatigue is present during games and training and that fatigue may increase the risk of injuries. As will be reviewed below, these risks of injuries can possibly be attributed to changes to muscles reflexes and also to kinetics and kinematics.

1.2.1 Fatigue and muscle reflexes

Fatigue is indicated by a reduction of maximal force or power output associated with sustained exercise and is reflected by a decline in performance (Rahnama, Reilly, Lees, & Graham-Smith, 2003). Because of this reduction in strength, fatigue has been hypothesized and shown to alter neuromuscular and biomechanical function (Melnyk & Gollhofer, 2007; Moore, Drouin, Gansneder, & Shultz, 2002; Wojtys, Wylie, & Huston, 1996). More specifically, these studies have looked at the effect of fatigue on hamstrings reflexes and tibial translation.

Wojtys et al. (1996) concluded that quadriceps and hamstrings fatigue which was induced on an isokinetic dynamometer, causes a slowing in reflex muscle responses in response to mechanically induced anterior tibial translation. More specifically, it was reported that combined fatigue of hamstrings and quadriceps muscles decreases the muscle reaction time and also the number of responses of both muscles. Similarly, Melnyk and Gollhofer (2007), after applying a fatigue protocol with the use of an isokinetic dynamometer on the knee flexors, found that the muscular response to mechanically induced anterior translation of the tibia was modified. Indeed, there was a significant decrease in reflex size in both components of the reflex response (short latency response and medium latency response) and a delay in reflex time of the hamstrings. Both of these research groups concluded that these alterations could possibly lead to an increased risk of sustaining musculoskeletal injury. The effect of fatigue on reflex response timing and amplitude was also studied by Moore and his group on both males and females (Moore, et al., 2002). In contrast to the previous researchers who studied mechanically induced anterior tibial translation (Melnyk & Gollhofer, 2007; Wojtys, et al., 1996), Moore and colleagues measured force and EMG activity of vastus lateralis after a tap on the patellar tendon had elicited the patellar tendon reflex (Moore, et al., 2002). Fatigue of the quadriceps was induced with continuous isokinetic contractions on a dynamometer. They concluded that just after the fatigue protocol, males showed a significant increase in EMG amplitude and a non-significant increase in force production of the quadriceps and females showed a non-significant decrease in EMG amplitude and a significant decrease in force production of the same muscles. According to the authors (Moore, et al., 2002), the causes of these noted gender differences remain unclear. It is possible that females were working close to their maximum power generation, leaving no room for an increase in EMG post-fatigue.

Through the research on the effect of fatigue on reflex time and amplitude, Melnyk and Gollhofer (2007) and Wojtys et al. (1996) also reported an increase in anterior tibial translation in response to mechanical perturbation. This increase, according to Wojtys and colleagues (Wojtys, et al., 1996), is due to the viscoelastic changes in the collagenous tissues of the knee and the fatigued knee stabilizers. According to Melnyk and Gollhofer (2007), the increase of the anterior tibial translation is due to the decrease in reflex size in both components of the reflex response of the hamstrings after fatigue. Both these studies agree on the fact that this increase in anterior tibial translation is associated with a mechanical loss of knee stability and increased risk of knee injuries.

1.2.2 Fatigue, kinematics, and risks of injuries

Many studies have specifically addressed the effect of muscle fatigue on lower limb kinematics and how these are altered in a manner that increases risk of injuries during dynamic activities such as running and jumping. Various fatigue protocols were used to study changes in EMG and kinematics during these activities.

A study on soccer players showed a predisposition of hamstring strain injuries during the latter stages of a protocol developed to replicate the physiological and mechanical demands made during soccer games (Small, McNaughton, Greig, Lohkamp, & Lovell, 2009). The researchers, reached this conclusion after measuring sprint times and lower-limb kinematics of participants at eight different points in time, during this 90-minute, multidirectional, soccer-specific fatigue protocol. After applying the protocol on nine healthy male semi-professional soccer players, they showed that fatigue causes time dependent alterations in sprinting kinematics such as decreased maximum combined hip flexion and knee extension angle. These results supported the hypothesis that the length of hamstrings becomes shorter after fatigue, which may cause increased predisposition to hamstring strain injury. Another interesting finding of this group is that they observed an increase in lower limb velocity after fatigue, which possibly impairs the ability of the hamstrings to decelerate the limb effectively to avoid risk of injury.

The effect of fatigue on sagittal plane lower limb joint kinematics during running was also tested by another research group (Kellis & Liassou, 2009). Fatigue of the sagittal plane lower-limb movers was induced using an isokinetic dynamometer. They studied the effect of fatigue on kinematic and electromyographic data of first the ankle plantarflexors and dorsiflexors, and then the knee extensors and flexors during running. They found that after the ankle musculature fatigue, the dorsiflexion angle at the initial impact phase of running was decreased, whereas after knee musculature fatigue, the knee flexion angle during toe-off was increased. The increased knee flexion at toe-off indicates that participants preferred to flex their knee more to achieve propulsion. Such adaptations show that participants have to change lower limb kinematics to compensate for muscle fatigue, possibly increasing the risk for injury. Moreover, EMG data revealed that ankle musculature fatigue caused a decline of biceps femoris EMG activity during the swing phase of running. As observed by Small et al. (2009), a decrease in EMG activity of biceps femoris during the swing phase could lead to a decrease in the ability for deceleration of knee extension. Again, these alterations could increase the risk of injuries. The authors also believe that this combined decline in muscle function and the alterations in kinematic data, following fatigue, is linked to impaired sprinting performance.

The effect of muscular fatigue on jumps has also been studied. In all of these studies, changes in either the kinematics or the EMG activity of muscles were observed that could increase the risk of injuries (Chappell, et al., 2005; McLean, et al., 2007; Orishimo & Kremenic, 2006; Padua, et al., 2006). The researchers used a variety of protocols which, through the performance of jump tasks, step-ups, squats and sprinting, induced fatigue to the lower-limb muscles of their participants. Chappell and his group (2005) found a decreased knee flexion angle during the landing phase of three different stop-jumps after fatigue, in contrast with Orishimo et al. (2006) and Padua et al. (2006) who found an increased knee flexion angle during landing. This difference could be due to the fact that different types of jumps were analyzed; indeed Padua et al. and Orishimo et al. did their measurements during hopping and Chappell et al. during stop-jumps. Kinematic data also showed that fatigue increases peak proximal tibial anterior shear force (Chappell, et al., 2005) and also peak knee internal rotation (McLean, et al., 2007) during the landing phase of the jumps; both groups of researchers feel that their findings are responsible for the increased risk of injuries. EMG data, on the other hand, showed that after muscular fatigue, individuals showed greater reliance on the ankle musculature than on the knee musculature (Padua, et al., 2006). Also, at the knee level, fatigue caused a 25% decrease in peak hamstring activation, leading to a more quadriceps dominant strategy, which can put the knee at increased risk of injury. Some of these studies also looked into the effect of gender on changes that occur following muscular fatigue. Women had increased peak knee abduction moments (McLean, et al., 2007) and also the effect of fatigue caused either more prominent changes in the quadriceps dominant strategy (Padua, et al., 2006) or an even higher peak proximal tibial anterior force (Chappell, et al., 2005), compared to men. Findings like those may explain why women are more susceptible to anterior cruciate ligament (ACL) injuries during fatigue, than men.

Ortiz et al (Ortiz, et al., 2010) tested the assumption that fatigue leads to alterations in knee muscle activation, peak knee joint angles and peak knee internal moments during drop jumps and hops, using an anaerobic protocol. These alterations can lead to impaired joint stability which can be responsible for lower limb injuries. They applied the Wingate fatigue protocol, which is an anaerobic protocol consisting of cycling bouts, in order to induce metabolic fatigue, to 15 physically active and healthy individuals; they then gathered kinematic, kinetic and electromyographic data in the lower limbs during two different jump tasks: a single-legged drop jump from a 40-cm box and a 20-cm, up-down, repeated hop task. What they found was a reduction in mean knee flexion angle during the fatigued session, which made them conclude that fatigue places the knee in an unstable situation. Some non-significant trends that were noticed in the study included an increase in the knee valgus angle, effectively placing the knee in an unstable position and possibly increasing the risk of knee injuries. EMG data however, did not yield any significant change between the pre-fatigued and the post-fatigued condition.

The effect of fatigue on kinematic properties during jumps has also been studied using protocols that induced endurance fatigue in individuals. Indeed, Moran and his group gathered kinematic data while healthy individuals performed drop jumps from different heights, before and after an endurance running fatigue protocol (Moran, et al., 2009). The running protocol was done on a treadmill and induced whole body endurance fatigue. The drop jumps were performed from different heights (15, 30, and 45 cm). They found that fatigue did not have an effect on joint angular kinematics but it did have an effect on tibial peak impact acceleration. More specifically, it was found that tibial accelerations at foot contact were larger during drop jumps after fatigue. They concluded that the increase in impact accelerations caused by fatigue may increase the risk of injury.

Fatigue is also associated with increased risk of injuries not only to lower-limbs, but also to other body parts. Indeed, an interesting study on the effect of fatigue on the dynamic stability of the torso and consequently on risk of upper body injuries was made by Granata and Gottipati (Granata & Gottipati, 2008). They gathered trunk kinematic data of 10 healthy individuals, while they were performing trunk extension and flexion exercises, before and after a trunk extension exercise fatigue protocol. They found that the torso was less stable after the fatiguing protocol than during unfatigued measurements. Their conclusions were based on the fact that the Lyapunov exponent (λ_{max}), which represents the rate at which kinematic disturbances change with time, was increased with fatigue in this experiment. They believe that these modifications contribute to disturbance perturbations that precipitate unstable intervertebral movement. Their results demonstrate that fatigue of the trunk extensor muscles impairs the neuromuscular stabilising control of dynamic torso movements and subsequently may increase the risk of tissue strain injuries.

After reviewing several studies, it is concluded that fatigue not only affects force production but also increases the possibility of injury during dynamic activities such as running and jumping. This increased risk of injuries is explained by the reported changes in upper body and lower-limb kinematics and EMG activity resulting from fatigue. It has been shown that fatigue increases tibial acceleration during jumps, increases proximal tibial anterior shear force and also leads to a more quadriceps dominant strategies. Moreover, comparison between genders showed that women, after fatigue, had more prominent changes in the quadriceps dominant strategy or an even higher peak proximal tibial anterior force, compared to men.

1.3 Muscle fatigue and proprioception

Not only does fatigue lead to a decrease in force production, it has also been shown to have a deteriorating effect on proprioception. Proprioception is the ability to detect, without the visual input, the spatial information and/or the movement of the limbs in relation to the rest of the body (Hogervorst & Brand, 1998).

1.3.1 Anatomical elements of proprioception

Sensory information is sent from mechanoreceptors to the central nervous system (CNS), contributing to stability and motor control (Hogervorst & Brand, 1998). These specialised mechanoreceptors are found in muscles, tendons, capsules, ligaments and skin throughout the body. It is generally accepted that the greatest contribution to joint position sense and kinesthesia is from muscle receptors, mainly muscle spindles and Golgi tendon organs (Eklund, 1972). The muscle spindles are sensory receptors, which are distributed throughout the belly of the muscles and send information to the nervous system about muscle length or rate of change of length. They are used to determine joint angulation in mid ranges of motion and are important in helping control of movement. The Golgi tendon organs are encapsulated sensory receptors through which muscle tendon fibers pass. They are located in the tendons of the muscles and transmit information about tendon tension or rate of change of tension (Guyton & Hall, 2006). Aside from these main receptors, there are also others receptors located in different structures. For example, the ACL has been shown to contain different types of mechanoreceptors (Hogervorst & Brand, 1998). These elements are important for proprioception and they can influence the activity of the muscles around the joint and the precision of the joint movements (Schutte, Dabezies, Zimny, & Happel, 1987). Because of the rich concentration of mechanoreceptors in the ACL, any rupture of it can lead to a lack of sensory input (Hogervorst & Brand, 1998; Kennedy, Alexander, & Hayes, 1982).

1.3.2 Evaluating proprioception

In order to be able to understand the studies about proprioception, we must know how it is quantified by the researchers. A review by Hiemstra stated that proprioception can be divided into two elements: joint position sensing (JPS) and sense of limb movement (kinaesthesia);the latter is observed by measuring the threshold to detection of passive motion (TTDPM) (Hiemstra, Lo, & Fowler, 2001). These two elements are commonly used as experimental outcomes by researchers to study proprioception and the effect of different variables (injuries, fatigue, etc.) on proprioception. Although performed using different procedures, both JPS and TTDPM are measured using the same apparatus (Fig.1.1).

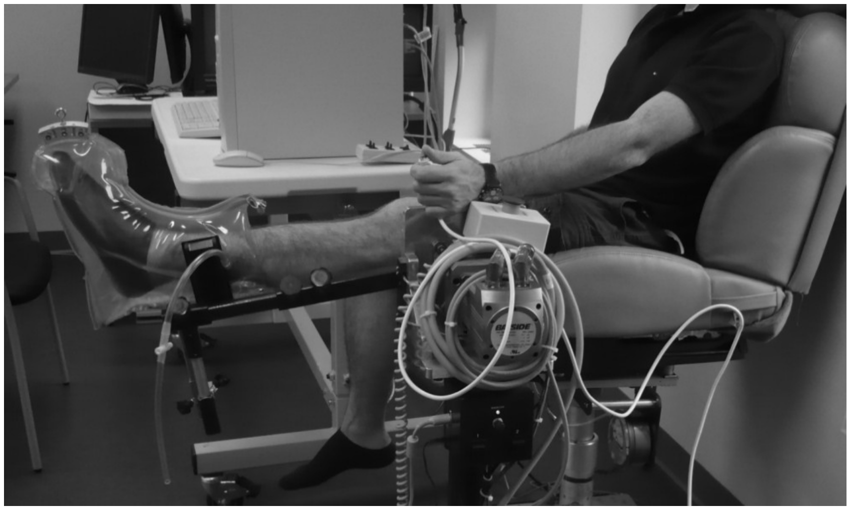


Fig. 1.1 JPS and TTDPM experimental setting. The individual is strapped on the dynamometer, with no visual and hearing stimuli and with an air cushion fitted around the leg. For TTDPM testing, participants had to use the stop-button.

JPS is usually evaluated on an isokinetic dynamometer and involves passive positioning and active repositioning (passive-active test). The individual sits on the dynamometer, with legs hanging freely, eyes closed to remove any visual input, wearing headphones to eliminate any hearing input and fitted with an air cushion around the leg, in order to neutralize cutaneous sensations. The tested extremity is passively moved, by the researcher, to the testing position, is held at this position for a short time period (usually a few seconds) and then repositioned in the initial position. Then the individual is asked to actively reposition the extremity to the testing position (Fig. 1.1). The difference in degrees between the testing and the repositioned positions is documented and represents JPS.

TTDPM is also evaluated on an isokinetic dynamometer. Again, the individual is seated on the dynamometer, while visual and hearing inputs are removed and an air cushion is placed around the leg. Then, the tested extremity is strapped on the dynamometer. After that, the device is slowly moved (usually slower than 1˚/sec) so that the knee is either flexed or extended and the participant is asked to press the handheld button when feeling a sensation of movement or a change in the knee starting position. TTDPM is defined as the angular change between the starting position and the position reached when the participant starts feeling the movement.

1.3.3 Effect of fatigue on JPS

JPS has been shown to be diminished in healthy individuals, after a fatigue protocol, which targeted only one muscle group (Mohammadi & Roozdar, 2010). Indeed, Mohammadi and Roozdar found that ankle JPS was decreased after fatiguing the ankle evertors using isometric contractions. JPS has also been shown to be diminished in healthy individuals following a fatigue procedure targeting different muscle groups simultaneously (Ribeiro, Mota, & Oliveira, 2007). The study was done on older individuals (age range 62-77) and the protocol induced fatigue of the knee extensors and flexors, simultaneously. The protocol was executed on an isokinetic dynamometer and the muscles were contracted in a concentric way. It was shown that there was a significant increase in absolute angular error, meaning that local muscle fatigue of the knee muscles significantly changed JPS.

Global lower-limb muscle fatigue has also been shown to affect JPS in healthy individuals (Lattanzio, Petrella, Sproule, & Fowler, 1997; Miura, et al., 2004; Skinner, Wyatt, Hodgdon, Conard, & Barrack, 1986). Two groups induced global lower-limb fatigue on their participants by applying running protocols (Miura, et al., 2004; Skinner, et al., 1986). Miura et al. (2004) found that a general load fatigue protocol consisting of running on a treadmill deteriorates knee JPS. After a similar fatigue sprinting protocol, Skinner and his research group (1986) found that participants did not reproduce position of the knee as precisely as before the application of the protocol. They concluded that this decrease in reproduction ability is due to the loss of efficiency of muscle receptors. Lattanzio et al. (1997), also observed a decrease in JPS, using a different fatigue protocol; participants cycled instead of running. Fatigue was induced in the participants by asking them to cycle on a cycle ergometer until maximal exhaustion. It is interesting to note that, in contrast with the previous mentioned studies, Lattanzio measured proprioception during weight-bearing conditions. Indeed, they measured knee proprioception in the standing position (closed kinetic chain). They asked the participants to stand on both feet with an electrogoniometer attached to the dominant leg. Participants flexed both knees, while standing, to a predetermined test angle. Then they returned to their initial stance and after a short interval they were asked to reproduce the test angle., In a closed kinetic chain position, as in the sitting position, researchers observed a reduction in knee proprioception, after the application of the fatigue protocol.

1.3.4 Effect of fatigue on TTDPM

Although researchers who used JPS to evaluate proprioception generally agree that fatigue diminishes proprioception, researchers who used TTDPM to study the effect of muscle fatigue on proprioception in healthy individuals reported contradictory results. Rozzi observed, after asking the participants to perform maximal effort knee flexion and extension exercises on a dynamometer, that combined fatigue of the lower–limb flexors and extensors led to increased TTDPM in healthy individuals (Rozzi, Lephart, & Fu, 1999b). Note that the significant increase in TTDPM was observed only during the extension movement, but not during flexion. However, Skinner et al. (1986) did not observe fatigue, induced by a sprinting protocol targeting both knee flexors and knee extensors, to affect TTDPM, neither in the extension nor the flexion direction.

Using both JPS and TTDPM, researchers have concluded that muscle fatigue decreases proprioception. This reduction in knee proprioception is hypothesized to be due to the decrease in muscle and joint receptor activity (Lattanzio, et al., 1997; Skinner, et al., 1986). According to Skinner's group this decrease in receptor activity could be due to reduced efficiency of the receptors. Lattanzio's group also suggested that fatigue decreases concentration levels which after exercise-induced fatigue can affect the ability to reproduce the knee angles. Thus, the observed decrease in knee proprioception might be due not only to muscular fatigue but also to lower concentration levels. This is another approach on how muscle fatigue affects, and specifically decreases, proprioception.

1.4 Muscle fatigue and balance

Colby (Colby, Hintermeister, Torry, & Steadman, 1999) have introduced an interesting description of the term "dynamic stability". They mention that "dynamic stability is the joint stability which is achieved through muscle coordination, proprioception and the ability to stabilize the knee joint, while the movement occurs". Therefore, proprioception is, at least in part, responsible for the ability of the body to maintain its dynamic stability. Since muscle fatigue diminishes the amount of force that can be generated, as well as proprioceptive ability, it is possible that muscle fatigue also affects balance. Indeed it has been proven that fatigue is one of the factors affecting lower limb dynamic joint stability during athletic tasks (McLean, et al., 2007). Most of the studies on the effect of fatigue on balance measured sway while standing either on one or on both legs, using center of pressure (COP), and sometimes the center of mass (COM), as outcome variables. In general, they reported an increase in sway in healthy participants after fatigue protocols, targeting several different muscle groups.

1.4.1 Effect of fatigue on double-limb stance

Many researchers studied the effect of muscle fatigue on balance during double-limb stance (quiet standing). Most of them focused on the ankle musculature and how fatiguing these muscles affects balance. Mello and his group studied young, healthy subjects (mean age 23.2 ± 3.6 years) and found that local fatigue of the plantarflexors, induced by sustained isometric plantar flexion, caused significant increases in the measured stabilometric data (Mello, Oliveira, & Nadal, 2007). More specifically, they found that body sway tends to be increased with fatigue, as confirmed by the increased area and velocity of the COP displacement. The fact that the postural sway is significantly affected by fatigue of the plantarflexors is also supported by Corbeil et al. (2003) who found that muscular fatigue of the ankle plantarflexors yielded a significant increase to the COP speed in a healthy group of participants (Corbeil, Blouin, Begin, Nougier, & Teasdale, 2003). Mello et al. (2007) also found that fatigue caused delayed recruitment of plantarflexors relative to changes in direction of COP displacements in standing. Recruitment of plantarflexors is known to be important for the correction of COP displacements, and the fact that the effect of muscle action over COP displacement is delayed could explain why body sway is increased with fatigue (Mello, et al., 2007).

Fatigue of other muscles such as lumbar extensors has also been shown to have a deteriorating effect on postural sway and balance during double-legged stance (Davidson, Madigan, & Nussbaum, 2004; Madigan, Davidson, & Nussbaum, 2006; Pline, Madigan, & Nussbaum, 2006). For example, COP velocity, displacement, and area have been shown to increase after fatiguing the lumbar extensor muscles with multiple back extension exercises (Davidson, et al., 2004; Pline, et al., 2006). Using a similar fatigue protocol, Madigan et al. (2006) found modifications in the COP and also of COM variables during stance in physically active males. They initially noted that, after fatigue, the standard deviation of COP velocity was slightly higher than the standard deviation of COM velocity in both the antero-posterior and medio-lateral planes. According to them, this finding wasn't surprising because the COP must oscillate more than the COM to keep the COM within the base of support. They also found the position of COM and COP to be moved anteriorly with fatigue and the standard deviation of COP and COM velocity to be significant increased. These changes in sway involved increases in joint angle variability and angular velocities, mainly to joints located closer to the fatigued site. According to the authors, this forward lean causes an increase in plantarflexor muscle activity that increases ankle stiffness, a necessary factor for controlling sway. However, they mentioned that their results showed that global measures of sway based on COM and COP are not necessarily indicative of the changes in individual joint kinematics and concomitant changes caused by the existence of fatigue.

An interesting study on the effect of fatigue of the neck muscles on balance was conducted by Gosselin and his group (Gosselin, Rassoulian, & Brown, 2004). They measured the amount of COP displacement, and its velocity along with the EMG activity of the cervical extensors in healthy, young participants during double-legged stance, before and after implementing a neck extensors fatigue protocol. Their fatigue protocol consisted of neck extensor isometric contractions. They noticed that after the fatigue protocol, COP displacements were significantly increased and, as a matter of fact, they concluded that the posturographic parameters that were affected showed similar patterns to parameters observed in patients who have suffered a whiplash injury. Their EMG data revealed that the cervical muscle EMG median frequency spectrum was shifted towards the lower frequencies after fatigue. This shift towards the lower frequencies was expected, since muscle fatigue is known to lead to a decline in median frequency (Mannion & Dolan, 1996).

Another interesting aspect has been found by Vuillerme's group, who measured the COP displacements under each foot separately during bipedal stance (Vuillerme & Boisgontier, 2010; Vuillerme, Sporbert, & Pinsault, 2009). After an isokinetic fatigue protocol that targeted either the hip abductors (Vuillerme, et al., 2009) or the plantarflexors (Vuillerme & Boisgontier, 2010) of the dominant leg, they found that the displacements were much greater under the non-fatigued foot compared to the fatigued foot during double-legged stance. This finding was attributed to the adaptive process of the body to cope with the impaired ability of the fatigued leg to control posture efficiently.

Pline and his colleagues not only looked at the effect of fatigue of lumbar extensor muscles on balance during double-legged stance, but were also interested in the effect of the duration and the workload of the fatigue protocol on postural sway (Pline, et al., 2006). After applying an extended lumbar extensors fatigue protocol, they measured three COP-based measures of postural sway: mean velocity, peak velocity, and sway area. They not only showed that fatigue increases postural sway, but also the effect that duration and work load had on postural sway. When fatiguing the participants at a lower workload over a longer period of time, larger increases in sway were elicited as compared to fatigue from a higher workload over shorter periods of time. They suggested that the lumbar muscle creep phenomenon is associated with impaired trunk control, which causes the increased body sway when experiencing workloads over a long period of time.

1.4.2 Effect of fatigue on single-limb stance

Some researchers also studied the effect of fatigue on balance in single-legged stance. As found with double-legged stance, many researchers used fatigue protocols that targeted the ankle musculature and most of them agreed that fatigue of these muscles increases COP displacement and velocity (Dickin & Doan, 2008; Ochsendorf, Mattacola, & Arnold, 2000; Springer & Pincivero, 2009; Suponitsky, Verbitsky, Peled, & Mizrahi, 2008; Vuillerme, Nougier, & Prieur, 2001; Yaggie & McGregor, 2002).

In the sagittal plane, several researchers have studied the effect of fatigue of the plantar- and dorsi-flexors on postural sway. After fatigue was induced, all of them observed an increase in postural sway, as shown by the increase in COP displacement or velocity (Dickin & Doan, 2008; Ochsendorf, et al., 2000; Springer & Pincivero, 2009). Note that the findings were found using different fatigue protocols. For example, Springer and Pincivero fatigued their participants with a series of maximal effort plantar and dorsi-flexion contractions while the two other groups fatigued the dorsiflexors and the plantarflexors of their participants simultaneously on an isokinetic dynamometer. Dickin et al. (2008) not only induced fatigue of the ankle muscles but also induced fatigue in the knee flexors and extensors through an isokinetic protocol. Fatigue of the knee muscles also resulted in increased COP displacements in the antero-posterior direction. Ochsendorf, looking not only at the effect of fatigue, but also at how orthotics influence the effect of fatigue, proved that orthotics may be an effective way of decreasing postural sway after an isokinetic fatigue protocol. These researchers found that the displacements of the COP were significantly less for the orthotic condition, both pre-fatigue and post-fatigue than for the non-orthotic post-fatigue condition.

Sway has also been studied after fatigue protocols that targeted either plantarflexors or dorsiflexors alone. In contrast to previously mentioned studies, Suponitsky et al. (2008), applied a protocol only in the dorsiflexion direction, inducing fatigue of the tibialis anterior in a quasi-isometric manner. They found that the amplitude of postural sway, which was quantified by measuring COP displacements, was significantly increased in the post-loading single-leg standing trials compared to the pre-loading trials. Researchers, who studied the effect of fatigue after a protocol which targeted only the plantarflexors, found contradictive results. Although Vuillerme et al. (2001) found that fatigue of the calf muscles increases COP displacement velocity, Adlerton and Moritz did not find fatigue of the calf muscles to affect these same variables (Adlerton & Moritz, 1996). We can hypothesize that this contrast is due to the different fatigue protocols used by the researchers. Although both on tiptoes, in Vuillerme's study, participants had to maintain an isometric contraction of their calf muscles by standing still on their tiptoes and in Adlerton's study, fatigue was induced by having their subjects repeatedly rise on their tiptoes of one foot until exhaustion. Adlerton and Moritz (1996) attributed their findings to either the increased reflex activity in muscles spindles or the increased muscle stiffness due to muscle fatigue, factors that contribute to maintenance of postural control.

By contrast, fatiguing the ankle invertors and evertors doesn't cause significant changes in COP displacements in either the sagittal or the frontal planes (Gribble & Hertel, 2004). The researchers attributed this finding to the ankle invertors and evertors not being involved in keeping balance during stance. However, combined fatigue of the plantarflexors, the dorsi-flexors, the invertors and the evertors, leads to the increase of COP displacements during quiet, single-legged stance (Yaggie & McGregor, 2002). By combining these studies we can hypothesize that evertors/invertors do not play as important a role as the ankle muscles that work in the sagittal plane (plantar and dorsi flexors) in keeping body balance. It is interesting to note that Gribble et al. (2004), who also looked into the effect of hip abductors/adductors, did find that fatigue of the hip abductors/adductors caused significant change in COP. This finding was attributed to the significant role that the hip muscles play in both the frontal and the sagittal plane in controlling balance, whereas ankle invertors and evertors, as we hypothesize, do not seem to play a role in maintaining balance.

Research that has been done on the effect of whole body fatigue on sway parameters during single-legged stance, also showed increases in COP displacements or velocity after the application of these fatigue protocols (Dickin & Doan, 2008; Springer & Pincivero, 2009). Dickin and Doan (2008) implemented a protocol which targeted the whole lower extremity; this was done by asking the participants to repeat squat jumps, involving both eccentric and concentric actions. On the other hand, Springer and Pincivero (2009), used exercises on a rowing ergometer to induce whole body fatigue in their participants, not only involving lower limb muscles but also muscles from the upper body. Both of these groups observed increases in COP displacements in both the antero-posterior and medio-lateral planes. Dickin and Doan (2008) also reported an interesting finding about the time duration of the effects of fatigue. Their postural measures were recorded before fatigue, immediately after the fatigue protocol, 10 minutes post-fatigue, 20 minutes post-fatigue and 30 minutes post-fatigue. COP displacements were still affected even after 30 minutes post-fatigue outlining that muscular fatigue imposed a prolonged internal perturbation to postural control.

Muscles play a major role in maintaining dynamic stability. Indeed, by reviewing studies on the effect of fatigue on single-legged and double-legged stance, we can conclude that muscle fatigue has a deteriorating effect on balance. During double-legged stance, fatiguing ankle, knee, lumbar and even neck muscles caused an increase in COP displacements and velocity. The same effects are notable after fatigue during single-legged stance. It is interesting to note that during bipodal stance with only one leg fatigued, COP displacements were much greater under the non-fatigued leg. Another interesting finding is that fatigue of the neck muscles deriving from isometric contractions, can mimic the effects of a whiplash injury can have on COP displacements.

1.4.3 Role of vision

The effect of vision on postural control following lower limbs fatigue protocol has also been investigated. Most of the previously mentioned researchers performed their experiments while their participants had their eyes either open or closed throughout the procedure without comparing balance in different vision modes (eyes open/eyes closed). However, Corbeil et al. (2003) and Vuillerme et al. (2001) measured the effect of fatigue on balance in both eyes closed and eyes open conditions. Both the researchers induced fatigue of the calf muscles of the participants, but used different protocols. In Corbeil's study the fatigue protocol consisted of repeated plantarflexion of both ankles and in Vuillerme's study fatigue was induced through isometric contractions of the calf muscles. Both teams measured displacements and velocity of the center of pressure. Corbeil concluded that removing vision increased sway in a similar fashion in both the fatigued and non-fatigued conditions. This finding suggests that vision can't compensate for the existence of fatigue. In Vuillerme's study the participants started the experiment with or without vision and throughout the procedure vision was suppressed or reinserted accordingly. Similarly to Corbeil et al., Vuillerme et al. found that when vision is removed there is an increase in sway, whether muscle fatigue is present or not. However, Vuillerme's group also found that the insertion of vision during a trial compensates largely for the effect of fatigue. Corbeil attributed their finding to the fact that localized muscle fatigue of the ankle plantarflexors affects the motor output of the postural control system more than the sensory system and Vuillerme et al. concluded that the availability of vision allowed their participants to immediately cope with the destabilizing effect induced by muscular fatigue.

1.6. Summary

Muscular fatigue is a natural consequence of prolonged exercise; it can be either central or peripheral depending on which point of the pathway is affected, and both types of fatigue can lead to decreased force production. Muscle fatigue has also been proven to increase muscle reflex time and amplitude and researchers, using both JPS and TTDPM, have concluded that muscle fatigue decreases proprioception probably as a consequence of affecting several neural receptors. This reduction in knee proprioception is hypothesized to be due to the decrease in muscle receptor activity. Furthermore, in several studies it has been concluded that with muscle fatigue, both the unilateral and the bilateral stance of a person are affected and the risk of injuries is increased. Specifically, velocity and displacement of the COP position are altered for both single-legged and double-legged stance after the application of different fatigue protocols that target various parts of the body, including neck, trunk, hip, shank and ankle. This increased risk of injuries is explained by the alterations in upper body and lower-limb kinematics and EMG activity by fatigue. Although balance is increased during quiet standing after fatigue, we do not know how it is affected in other positions, such as during squats.

Chapter 2 – Rationale, Objectives and Hypotheses

2.1 Rationale

As expected, a search of the background literature has indicated that the stability of a fatigued person is significantly decreased during unilateral and bilateral stance. The finding that draws our attention is the fact that there are few studies that have investigated the effect of muscle fatigue during other positions. Participating in everyday activities or sports activities, quiet standing is not the only stance that the body adopts. Balance is also necessary during activities where the joints of the body are in different positions. For example, during static squats, joints of the lower limbs are flexed and the extensors are under greater stress, maintaining the position against gravity. The squat movement is commonly found in the routine of movements during different sports. For example, in soccer the goaltender is usually in a semi-squat position, in football and rugby the front-line defence men are in a semi-squat position while they are waiting to tackle the players of the opposite team. Moreover, in volleyball the players are in a semi-squat position while they are waiting to receive the serve by the opposite team and in wrestling the opponents usually keep a semi-squat position, in order to keep their center of mass lower and be more stable to deflect an opponent`s offensive move.

In addition, squats are a commonly closed-kinetic exercise, used broadly during the rehabilitation process of different musculoskeletal injuries, mainly after ACL-injuries or ACL-reconstructions. Weight-bearing exercises, like squats, are less stressful, similar to functional movements and safer than non-weight bearing movements (Levangie, 2011). This study could therefore eventually be extended to injured populations.

This is why we believe that measuring balance during squats will be useful and will help us draw conclusions about the biomechanical alterations that fatigue causes to the body during activities. Moreover, when the quadriceps are activated (such as during squats), they pull the tibia anteriorly which is problematic in injured populations, such as in ACL-injured patients. The findings of Chmielewski that there is a strong correlation between weight distribution during squats and quadriceps strength in an ACL-injured group (Chmielewski, Wilk, & Snyder-Mackler, 2002), make us believe that squats during fatigued situations may be more difficult to perform and also can be used to evaluate patients. Studying how fatigue affects weight distribution in healthy participants during squats is a starting point to make comparisons with injured populations. Therefore, by evaluating how weight distribution is affected after a muscular fatigue protocol, which resembles the fatigue that the athletes might feel during sports situations such as during a game, it may eventually be possible to use this technique as a screening procedure before deciding if any athlete, who had an injury, is ready to return to training and games. Also, by measuring the activity of specific leg muscles, it may be possible to see how muscle recruitment is different after muscle fatigue in healthy populations, so that possible compensations used by injured populations to control stability could eventually be described.

2.2 Objectives

The objectives are to define and compare, during static squats, COP position, weight distribution and activity of the muscles in the lower extremities of a healthy population before and after a fatigue protocol.

2.3 Hypotheses

Our first hypothesis is that significant differences in COP position and weight distribution, after fatigue, will occur. Specifically, we expect COP position and weight distribution in the sagittal plane to move forward, but expect no difference in the frontal plane. Our second hypothesis is that the activity of the lower limb muscles, after fatigue, will be different. For example we expect the quadriceps and the gastrocnemius to show an increase in their activity.

Chapter 3 – Manuscript

Muscular fatigue affects center of pressure position and lower-limb muscle activation during static squats in healthy individuals

Ioannis Makris, BSc, Patricia A. McKinley, PhD, Nancy St-Onge, PhD

A similar version of this manuscript will be submitted to the

Journal of Orthopaedic and Sports Physical Therapy

3.1 Abstract

Muscular fatigue is a natural consequence of prolonged exercise and has been shown to decrease force production. Muscle fatigue has also been proven to alter reflexes and to decrease proprioception, probably as a consequence of affecting several neural receptors. Furthermore, it has been reported that fatigue affects balance during unilateral and bilateral stance. Although balance is decreased during quiet standing after fatigue, it is not known how it is affected in other positions, such as during static squats. The objectives of our study were to define and compare, during squats, the weight distribution, COP position in the sagittal plane and the activity of the lower-limb muscles in healthy individuals before and after a fatigue protocol. Fifteen healthy, physically active individuals (age 18-30 years) participated in our study. Data were collected while the participants were in different static squatting positions, before and after a lower-limb muscle fatigue protocol. Pressure distribution was collected using a MatScan and was used to compute weight distribution in two planes (coronal and sagittal) and COP position in one plane (sagittal). Electromyographic (EMG) data from 14 lower-limb muscles (7 on each side) were collected. The fatigue procedure was successful in inducing perceived muscle fatigue as measured with the Borg scale. Our results show that weight distribution was not affected in either planes but COP position was transferred anteriorly after fatigue, especially at 60^{o} and 90^{o} knee flexion. Fatigue also increased the activity of RF, VM, BF, and GL, while deeper squatting positions increased the activity of RF, VM, ST, and TA. These changes may lead to increased risk of injury in athletes, when they become fatigued during practices or games, especially in deeper squat positions.

3.2 Introduction

"The term muscle fatigue is used to denote a transient decrease in the capacity of the muscle to perform physical actions" (Enoka & Duchateau, 2008). According to the authors, during fatigue, muscles decrease their ability to produce maximal force or power. Moreover, it has been shown, using both joint position sense (JPS) and threshold to detect passive motion (TTDPM), that fatigue affects proprioception. This reduction in proprioception may be due to a decrease in muscle receptor activity (Lattanzio, et al., 1997; Skinner, et al., 1986). Lattanzio's group also suggested that fatigue decreases concentration levels which can affect the ability to reproduce target knee angles after exercise-induced fatigue (Lattanzio, et al., 1997). JPS has been shown to be diminished in healthy individuals, after a fatigue protocol that targeted only one muscle group or a group of antagonist muscles (Mohammadi & Roozdar, 2010; Ribeiro, et al., 2007). Global lower-limb muscle fatigue induced by activities such as cycling, running on a treadmill, or sprinting, has also been shown to affect JPS in healthy individuals (Lattanzio, et al., 1997; Miura, et al., 2004; Skinner, et al., 1986). Although researchers who used JPS to evaluate proprioception generally agree that fatigue diminishes proprioception, researchers who used TTDPM to study the effect of muscle fatigue on proprioception in healthy individuals found contradictory results. Rozzi observed that combined fatigue of the lower–limb flexors and extensors led to increased TTDPM during the extension movement, but not during flexion (Rozzi, Lephart, & Fu, 1999a). However, Skinner did not observe fatigue of the hamstring musculature to affect TTDPM, neither in the extension nor the flexion direction (Skinner, et al., 1986). Difference in results may be due to the fact that Skinner's group induced fatigue only in the hamstrings in contrast to Rozzi's group who elected to fatigue both the extensors and the flexors of the knee.

Since muscle fatigue diminishes the amount of force that can be generated, as well as proprioceptive ability, it is possible that muscle fatigue also affects balance. Indeed, fatigue of different lower-limb muscles, lumbar extensors, and even neck muscles has been shown to lead to an increase in center of pressure (COP) displacements and velocity in healthy individuals during both single-legged and double-legged stance (Adlerton & Moritz, 1996; Corbeil, et al., 2003; Davidson, et al., 2004; Dickin & Doan, 2008; Gosselin, et al., 2004; Gribble & Hertel, 2004; Madigan, et al., 2006; Mello, et al., 2007; Ochsendorf, et al., 2000; Pline, et al., 2006; Springer & Pincivero, 2009; Suponitsky, et al., 2008; Vuillerme & Boisgontier, 2010; Vuillerme, et al., 2001; Vuillerme, et al., 2009; Yaggie & McGregor, 2002). It is interesting to note that Mello not only found an effect of muscle fatigue of the plantarflexors on COP in double-legged stance but also reported a delay in recruitment of the fatigued muscles (Mello, et al., 2007). According to these authors, the delayed recruitment of plantarflexors may be responsible for the increased COP displacements. An interesting finding has been reported by Vuillerme's group (Vuillerme & Boisgontier, 2010; Vuillerme, et al., 2009), who measured COP displacements under both feet during bipedal stance. They found that fatigue of either the hip abductors or the plantarflexors of the dominant leg led to greater COP displacements under the non-fatigued foot compared to the fatigued foot. This was attributed to the adaptive process of the body to cope with the impaired ability of the fatigued leg to control posture efficiently. Contradictory results were reported on the effect of fatigue of the calf muscles on COP displacement velocity during single-legged stance (Adlerton & Moritz, 1996; Vuillerme, et al., 2001). Indeed, in Vuillerme's study, COP displacement velocity was increased after fatigue of the calf muscles whereas in Adlerton's study, it was not affected. These contradictory findings may be due to the use of different fatigue protocols. In Vuillerme's study, participants had to maintain an isometric contraction of their calf muscles by standing still on their tiptoes and in Adlerton's study, fatigue was induced by having subjects repeatedly rise on their tiptoes until exhaustion. Adlerton and Moritz attributed their findings to the increased reflex activity in muscles spindles or the increased muscle stiffness due to muscle fatigue. We believe that these two factors may have permitted better control of the COP displacement velocity.

Although contradictory results on the effect of muscle fatigue have been found, most of the researchers agree that muscle fatigue decreases standing balance during both single-legged and double-legged stance. To our knowledge, the effect of fatigue on squats is not known. Our goal was therefore to define and compare, during squats, the COP position, the weight distribution and the activity of lower-limb muscles in a healthy population before and after a fatigue protocol. Our first hypothesis was that fatigue would affect COP position and weight distribution in the sagittal plane, but not in the coronal plane. Our second hypothesis was that activity of lower-limb extensor muscles would be different after the fatigue procedure.

3.3 Methods

3.3.1 Participants

Fifteen healthy, physically active (participating in resistance and aerobic training 1-4x per week) male participants (age: 22.5±2.67 yrs old, weight: 81.1±11.21 kg, height: 180.3±5.74 cm) volunteered for this project. All participants were right-leg dominant and none had a history of ankle, knee, hip and back injury or deformation. Also, participants did not have any visual, vestibular or neurological condition. Dominant leg was determined by asking the participant which leg they would kick a ball with. After the procedure was explained to the participants, they signed a consent form approved by the institutional ethics committees (*Centre de Recherche Interdisciplinaire en Réadaptation du Montréal Métropolitain* and Concordia University).

3.3.2 Preparation

Bipolar surface electrodes were positioned bilaterally on the following muscles, according to the SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) guidelines: rectus femoris (RF), vastus medialis (VM), semitendinosus (ST), biceps femoris (BF), tibialis anterior (TA) gastrocnemius lateralis (GL), and gastrocnemius medialis (GM). Electrode placement is detailed in Table 3.1. The reference electrode was positioned over the right patella. The skin was shaved and abraded with an alcohol pad to ensure the proper adhesion and better conductance between the electrodes and the skin. A 1-D electrogoniometer (Noraxon®, Phoenix, U.S.A.) was also fixed to the right thigh and leg with double-sided tape. One part of the goniometer was attached on the thigh on the line connecting the trochanter with the center of rotation of the knee joint and the other part was attached on the leg on the line connecting the center of rotation of the knee joint with the malleolus (see Fig. 3.1).

Table 3.1 EMG electrodes attachment sites

MUSCLES	ATTACHMENT SITE
RF	50% between anterior spina iliaca superior and superior part of the patella
VM	80% between anterior spina iliaca superior and anterior border of the medial ligament
ST	50% between ischial tuberosity and medial epicondyle of tibia
BF	50% between ischial tuberosity and lateral epicondyle of tibia
TA	1/3 of line connecting tip of fibula and tip of medial malleolus
GL	1/3 on line between the head of fibula and the heel
GM	on most prominent bulge of muscle

Fig. 3.1 Attachment of electrogoniometer on the right thigh and leg of one participant.

3.3.3 Experimental protocol and data acquisition

The EMG activity and electrogoniometer data were recorded with a TeleMyo 2400T G2 (Noraxon®, Phoenix, U.S.A.) with a sampling frequency of 1500 Hz. The recordings were filtered using a zero-lag 4th order Butterworth filter with a 10-350Hz bandpass before proceeding with the analysis. A Matscan (Tekscan®, Boston, U.S.A.) was used to record pressure distribution with a sampling frequency of 100 Hz per sensor. The MatScan has a sensing area measuring 435.9*368.8 mm and contains 2288 sensors, organized in 44 rows and 52 columns. The spatial resolution of the MatScan is 1.4 sensels/cm^2

The general protocol is outlined in Fig. 3.3. Participants were asked to stand on the pressure mat, with no footwear, eyes open and hands on the hips. They were instructed to keep their feet shoulder width apart and no specific instruction was given for trunk position (Fig. 3.2).

Throughout the experimental procedure distance between the feet, feet orientation and trunk orientation were not controlled in order for the participants to adopt a more "natural" stance throughout the experimental procedure. However, after the completion of the study, we used the MatScan data to compute the distance between the heels and the orientation of the feet and evaluate whether they were affected by fatigue. The angle between the feet line was not affected by fatigue. However, distance was slightly affected by fatigue (pre-fatigue mean distance: 28.8cm, post-fatigue mean distance: 28cm). Initially the participants had to stand straight on the MatScan. They then flexed their lower limb joints until the knees were flexed 30˚. Subsequently, they increased flexion until they reached 60˚ knee flexion and finally 90˚. They were then asked to return to the initial position. In each position (0˚, 30˚, 60˚, and 90˚), EMG and pressure distribution data were collected for a duration of 500 ms.

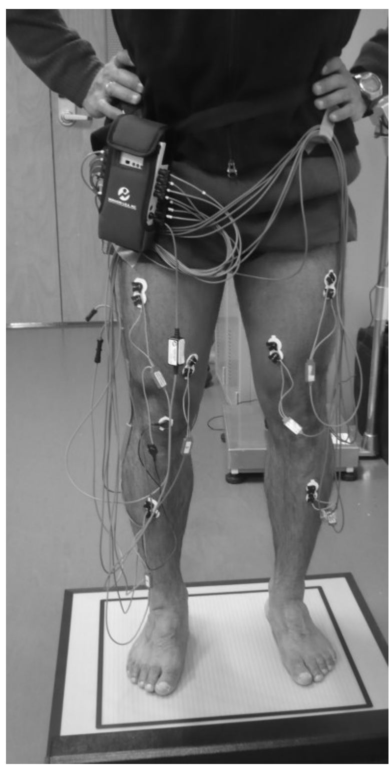


Fig. 3.2 Participant in the initial position. Feet are shoulder width apart and hands are on hips. Feet were not moved during squats so that position of the feet was the same for all positions.

This movement was repeated three times. The electrogoniometer data were displayed online and the information was used to guide the participant to the correct knee position. The participants then performed a fatigue protocol as outlined below. Immediately after performing the fatigue procedure, another set of three squats was performed and recorded. Perceived lower-limb muscle fatigue was assessed at four different times during the experimental procedure using the modified Borg scale (Borg, 1970): T1- before the first set of squats, T2- after the first set of squats, T3- after the fatigue procedure, and T4- after the second set of squats (Fig. 3.3). The Borg scale ranges from 0, representing no fatigue at all to 10 representing very, very strong fatigue. The Borg scale has often been used in research to evaluate perceived fatigue (Heuser & Pincivero, 2010; B. J. Taylor & Romer, 2008; Troiano, et al., 2008).

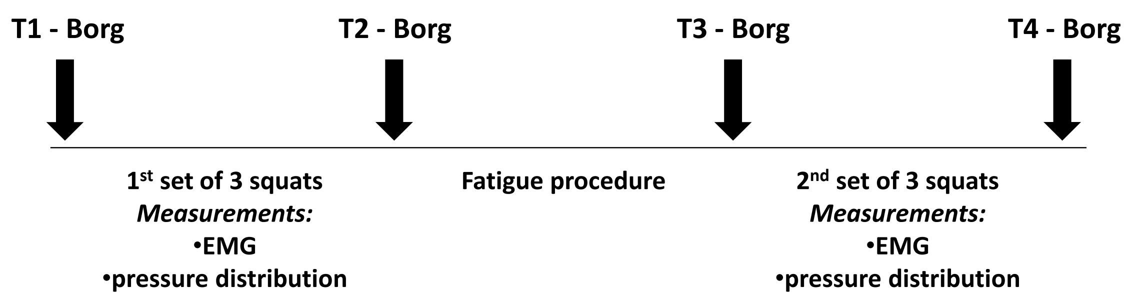


Fig. 3.3 Experimental procedure. During the testing procedure, participants performed two sets of three squats at each knee position, during which EMG and pressure distribution data were collected. Between these two sets, a fatigue procedure was performed. Fatigue was measured four times throughout the experiment using the Borg scale (T1, T2, T3, T4).

3.3.4 Fatigue procedure

A modified version of the fatigue protocol developed by Padua (Padua, et al., 2006) was used to fatigue the lower limb muscles. Participants were asked to perform squats from 0˚ to 60˚ of knee flexion while wearing a vest which contained 33% of their body weight. The squats were performed at a frequency of 50 squats per minute, with audio feedback provided by a metronome. Knee angle readings from the electrogoniometer were displayed on a computer screen as visual feedback to the participants. The fatigue protocol was continued until the participants fell four squat cycles behind the pace or failed to complete two successive squat cycles with the required amplitude. This type of fatigue protocol ensures that people with different fitness levels reach the same level of fatigue, regardless of training background and specific activities. Participants were constantly motivated by an experimenter to continue the procedure.

3.3.5 Data analysis

For each trial, root mean square (RMS) of the filtered signals was computed for each knee position (0°, 30°, 60°, and 90°) and each muscle. For each participant, results for the three pre-fatigue and the three post-fatigue trials were averaged in order to obtain one pre-fatigue and one post-fatigue value for each muscle and each knee position. The averaged RMS values for all pre-fatigue and post-fatigue positions were then normalized with the RMS value averaged across the three pre-fatigue trials at 90˚ of knee flexion for each muscle. EMG data is therefore presented in tables as a percentage of the amount of muscle activity at 90^{o} before the fatigue procedure (i.e., mV/mV at 90^{o} pre-fatigue).

Weight distribution was computed, for both the coronal (mediolateral body axis) and the sagittal (anteroposterior body axis) planes, using the MatScan data. The MatScan measurements were recorded with a frequency of 100 Hz for 500 ms per recording, providing 50 time frames with a duration of 50 ms each. For the coronal plane, the difference in percentage of weight under each foot was found for each time frame. These percentages were calculated by dividing the sum of the pressures under each foot by the sum of the pressures exerted on the MatScan. Data are therefore presented in percent body weight. The mean for all the frames was then computed for each recording. For each participant, these values were averaged for the three pre-fatigue and the three post-fatigue trials separately in order to obtain one pre-fatigue and one post-fatigue value in each of the four squatting positions (0°, 30°, 60°, and 90°).

For the sagittal plane, the feet were split into 20 sections distributed antero-posteriorly. The MatScan pressure mat contains 2288 sensors, organized into 44 rows from the front to the back and 52 columns (Fig. 3.4). For each trial, the mean of all 50 time frames was first computed for each knee position. The mean pressure of the columns for each row was then computed creating a vector containing 44 elements. The first and last rows with at least one sensor activated were used to delineate the feet, creating a new vector that could be of a different length for different trials and participants. This new vector was then mathematically transformed into a 20 element vector. This last vector was then normalized using the sum of all elements, creating a new 20 element vector expressed as a percentage. The 20 elements of the normalized vector add up to 100 and represent the antero-posterior weight distribution of both feet combined.

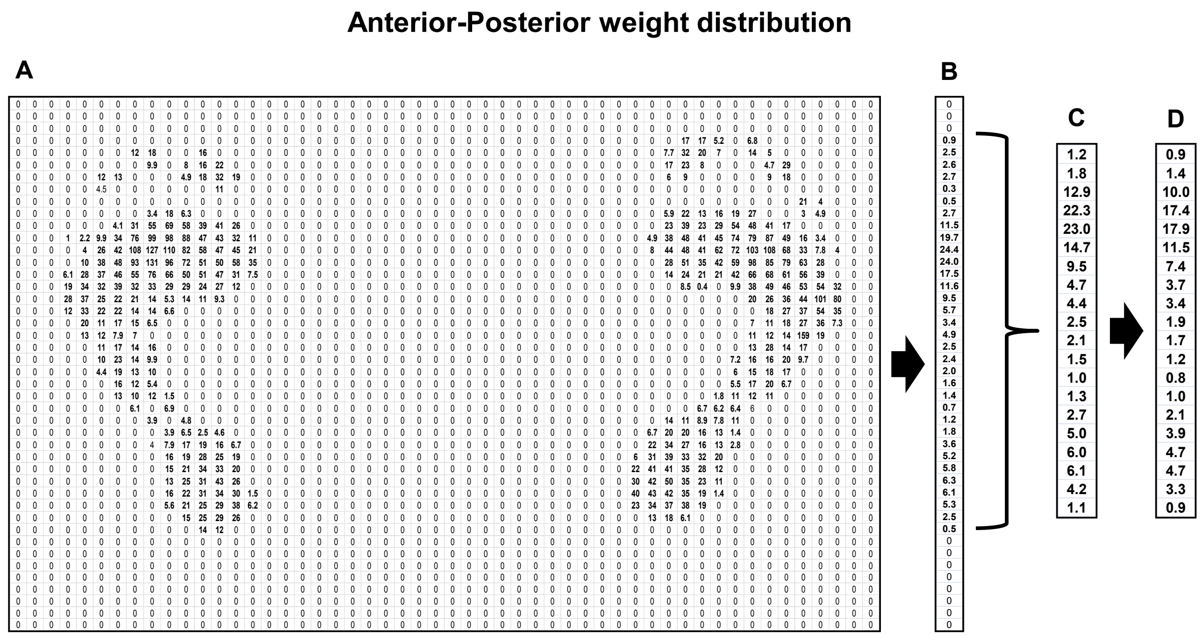


Fig. 3.4 Analysis of the MatScan data for weight distribution in the sagittal plane. A. Representation of the raw MatScan data for one trial. The numbers represent pressure on each of the individual sensors. B. Vector representing the mean pressure for each row of A. C. Mathematically converted vector B into a 20-element vector. D. Vector C normalized to body weight of the participant (values add up to 100).

For each participant, the normalized vectors were averaged for the three pre-fatigue and three post-fatigue trials in order to obtain one pre-fatigue and one post-fatigue vector for each of the four squatting positions. The 20 elements of each vector were then split into two parts: the 12 top elements for the anterior part and the bottom 8 elements for the posterior part. The separation of the 20 elements into 12 top elements and 8 bottom elements was done by visually inspecting the data and using the foot arch as a cutting point (Fig. 3.5). Two different analyses were then performed using the 12 anterior and the 8 posterior elements. For the first analysis, the peak values were obtained for both the anterior and the posterior parts on each vector. These values represent the pressure peaks on the anterior and posterior parts of the feet. This analysis provided two peak values (one for the anterior part and one for the posterior part) for each fatigue condition (pre and post) and each of the four knee positions for each participant. For the other analysis, the top 12 elements were added up, representing global pressure on the anterior part of the feet and the bottom eight elements were added up, representing global pressure on the posterior part of the feet. The sum of the 12 top elements was then subtracted from the sum of the eight bottom ones, creating the difference in percentage between the anterior and the posterior parts of the feet. There were therefore eight antero-posterior differences for each participant, one for each of the four knee positions, pre- and post-fatigue.

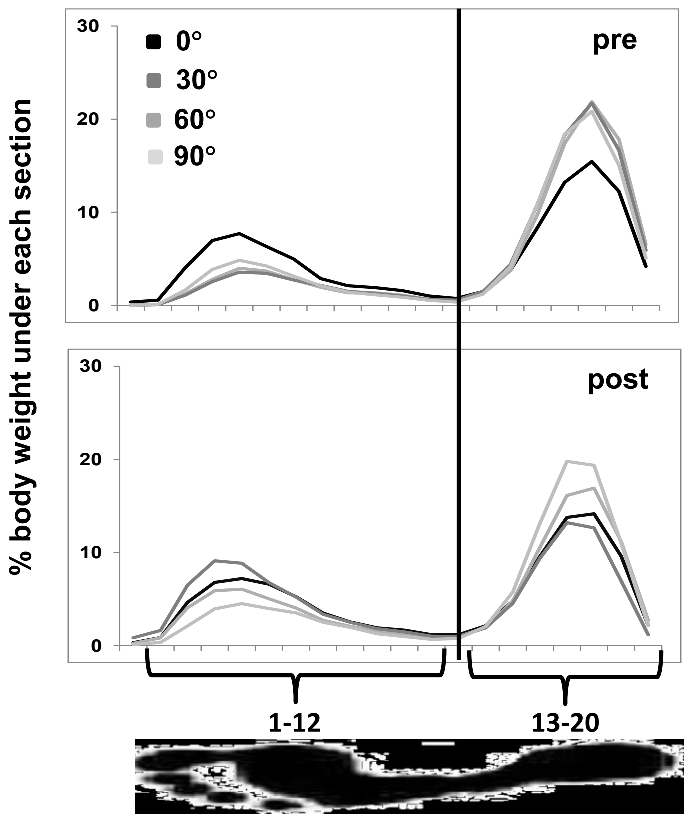


Fig. 3.5 Weight distribution data in the sagittal plane for one participant, for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), pre- and post-fatigue. The section with no (or little) weight was used as a cutting point between the anterior part and posterior part of the feet. This resulted in the 12 top elements being used to represent the anterior part of the feet and the 8 bottom elements being used to represent the posterior part of the feet.

COP position was computed, for the sagittal plane, using the MatScan data. For each trial, the mean of all 50 time frames was first computed for each knee position. The COP position was computed relative to the position of the heels using the bottom border of a box created around the feet (Fig. 3.6). We then used the height of the box to calculate and express the COP position as a percentage of the height of the base of support base. The height of the box was from the tip of the toes to the bottom of the heels. For each participant, the COP values were averaged for the three pre-fatigue and the three post-fatigue trials separately in order to obtain one pre-fatigue and one post-fatigue value in each of the four squatting positions (0°, 30°, 60°, and 90°).

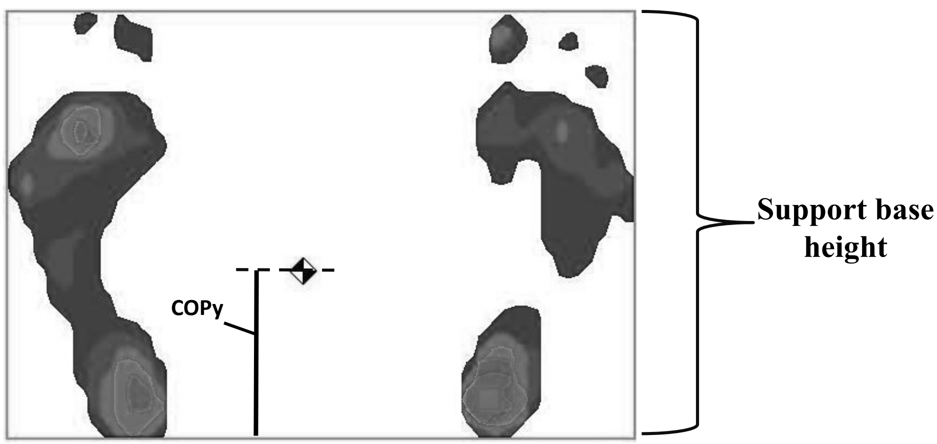


Fig. 3.6 COP position was computed relative to the position of the heels and expressed as a percentage of the height of the base of support. The diamond represents the COP position.

3.3.6 Statistical analyses

A repeated-measures analysis of variance (ANOVA) was performed to compare perceived muscle fatigue measures (Borg scale) assessed at four different times during the experiment (see Fig. 3.3 above). Note that data were missing for one of the participants and could therefore not be included in the analysis. Since standard deviations were large for weight distribution (both non-dominant - dominant and antero-posterior) and COP, data were first assessed for normality using the Shapiro-Wilk test and were shown to be normally distributed. For weight distribution in the coronal and the sagittal plane, a repeated-measures, two-way ANOVA with two within factors (fatigue (pre/post) * position (0˚, 30˚, 60˚, 90˚)) was performed. For the pressure peaks, a repeated-measures, three-way ANOVA with three within factors (fatigue (pre/post) * position (0˚, 30˚, 60˚, 90˚) * peaks (anterior/posterior)) was performed. For COP position in the sagittal plane, a repeated-measures, two-way ANOVA with two within factors (fatigue (pre/post) * position (0˚, 30˚, 60˚, 90˚)) was performed. For the EMG activity (RMS values), a repeated-measures, three-way ANOVA was used for each muscle, with three within factors (body side of the muscle (left/right) * fatigue (pre/post) * position (0˚, 30˚, 60˚, 90˚)). When a significant main effect was found, *post-hoc* analyses were performed using Tukey tests of comparisons. A significance level of $p<0.05$ was used for all analyses.

3.4 Results

3.4.1 Fatigue

Perceived lower-limb muscle fatigue was recorded at four different times during the experimental procedure: T1. before the first set of squats, T2. after the first set of squats, T3. after the fatigue procedure and T4. after the second set of squats (Fig. 3.7). Results show that the fatigue procedure was successful in increasing perceived muscle fatigue as measured with the Borg scale ($F(3,39)=65.8066$, $p<0.0001$). *Post-hoc* analysis revealed a significant difference between the pre-fatigue measures (T1: 0.18±0.54; T2: 0.89±1.06) and the post-fatigue measures (T3: 6.07±1.98; T4: 4.71±2.09). The pre-fatigue recordings were smaller than 1, meaning that the participants perceived their fatigue to be less than “very weak”. Immediately after the fatigue protocol (T3), the mean was close to 6, meaning that participants perceived their fatigue to be between "strong" and "very strong". The duration of the fatigue procedure lasted between 3 and 27 minutes depending on the participant. The time required to reach fatigue probably depends on the level of fitness but may also be dependent on the style of training the participants were involved in. After the second set of squats (T4), the mean was close to 5, meaning that fatigue was perceived to be "strong". There was also a significant difference between the post-fatigue measures, with T4 being less than T3.

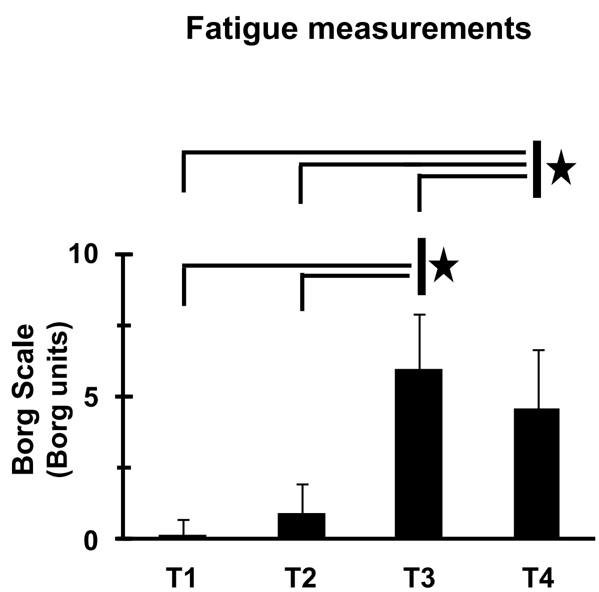


Figure 3.7 Borg Scale scores for perceived muscle fatigue, averaged across participants, at four different times throughout the experiment: T1. before the first set of squats (0.18±0.54), T2. after the first set of squats (0.89±1.06), T3. after the fatigue procedure (6.07±1.98), T4. after the second set of squats (4.71±2.09). The stars represent significant differences.

3.4.2 Weight distribution

Fig. 3.8 (left panel) shows weight distribution between feet (%non-dominant-%dominant) for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), before and after the fatigue protocol. Statistical analysis for the non-dominant - dominant weight distribution revealed no significant effect for either position or fatigue (position: ($F(3, 42)=0.1524$, $p=0.9275$; fatigue:

F(1,14)=2.6107, p=0.1284). There was also no significant position*fatigue interaction (F(3,42)=0.9859, p=0.4086).

Fig. 3.8 (right panel) shows weight distribution between the posterior and the anterior parts of feet (%posterior-%anterior) for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), before and after the fatigue protocol. Statistical analysis for the antero-posterior weight distribution revealed no significant effect for either position or fatigue (position: (F(3,42)=2.7342, p=0.0555; fatigue: F(1,14)=0.4977, p=0.4920). There was also no significant position*fatigue interaction (F(3,42)=1.7401, p=0.1734).

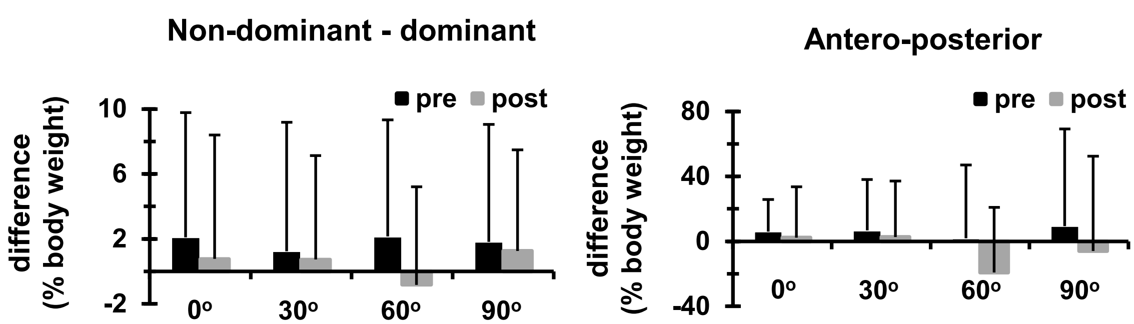


Figure 3.8 Weight distribution difference in the coronal and the sagittal plane averaged across participants. Data are shown for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), before and after the application of the fatigue protocol. The black bars represent pre-fatigue data and the grey bars represent post-fatigue data. The y axis represents the difference in percentage of body weight on each side computed by dividing pressure under each foot by the total pressure exerted on the Matscan.

Weight distribution on the 20 antero-posterior sections of the feet is displayed in figure 3.9. Data are shown for all four positions, before and after the fatigue protocol. In this figure, it can be seen that independent of the fatigue condition or knee position, two peaks are formed, one under the anterior part of the feet and one under the posterior part. Between these two peaks, there is a region with very low pressure which is positioned under the foot arch and where participants applied much less weight than under the ball of the feet or under the heel. In the pre-fatigue condition, weight distribution was almost the same for all the different knee positions, except at 90˚ for which there was a higher peak on the posterior part of the feet. In the post-fatigue condition, the anterior peaks were of similar amplitude for 0˚ and 30˚ and display lower values than at 60˚ and 90˚; the posterior peaks are of similar amplitude for 0˚, 30˚, and 90˚.

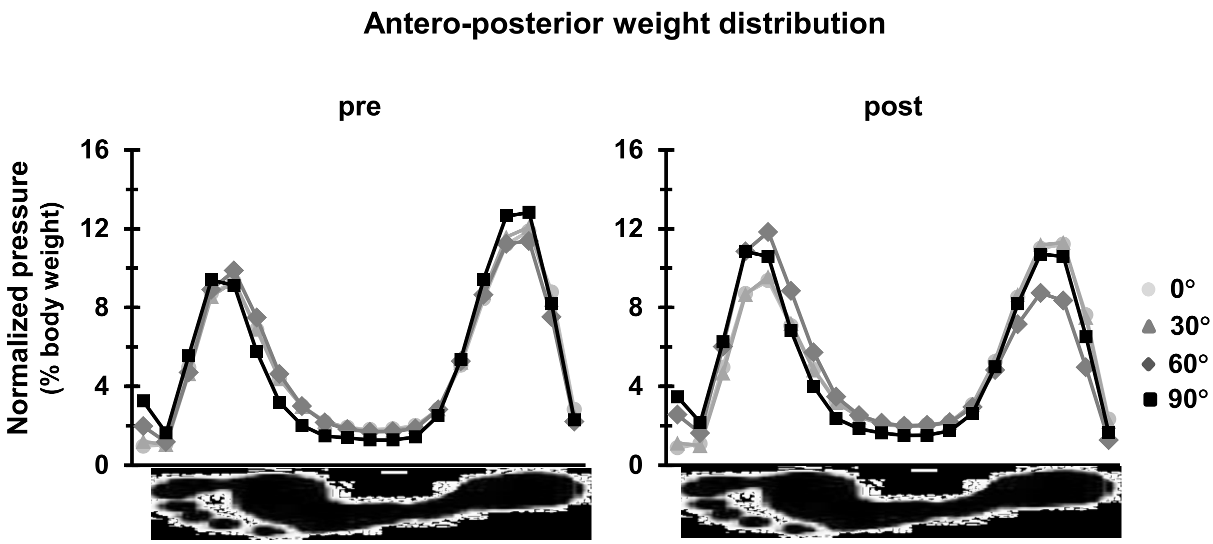


Fig. 3.9 Weight distribution in the sagittal plane averaged across participants. Feet were separated into 20 antero-posterior sections. Pressure under each section was normalized to body weight. Data are shown for the four different knee angle position (0˚, 30˚, 60˚, 90˚), before and after the application of the fatigue protocol.

The statistical analysis on the anterior and posterior peaks revealed a significant effect of knee position (F(3,42)=8.767, p=0.0001) with *post-hoc* analysis identifying higher peaks at 90°. There was also a significant effect of fatigue (F(1,14)=18.279, p=0.0007) with *post-hoc* analysis revealing lower peaks after the fatigue procedure. There was also a significant fatigue*peak interaction (F(1,14)=6.201, p=0.025), with the anterior peaks being higher after fatigue (Fig. 3.10). There was no position*peak or position*fatigue interaction for pressure peaks.

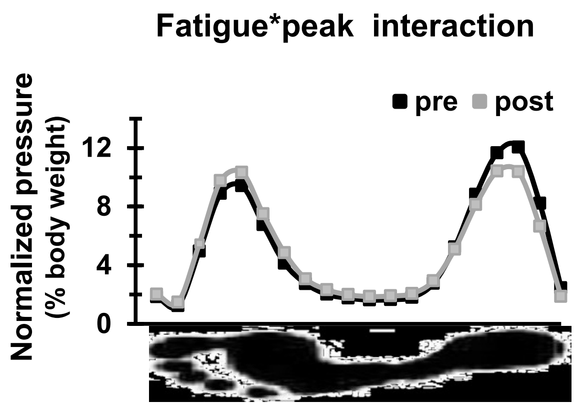


Fig. 3.10 Fatigue*peak interaction for weight distribution in the sagittal plane. The black line represents the data before fatigue and the grey line represents the data after fatigue.

3.4.3 COP

Fig. 3.11 shows COP position in the sagittal plane for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), before and after the fatigue protocol. Statistical analysis for the COP position revealed a significant effect for fatigue as well as a significant knee position*fatigue interaction (fatigue ($F(1,14)=5.1856$, $p=0.0389$; position*fatigue ($F(3,42)=4.9887$, $p=0.0047$)) with fatigue having an effect mainly at 60^o and 90^o. On the other hand, there was no significant effect for knee position ($F(3,42)=0.7201$, $p=0.5456$).

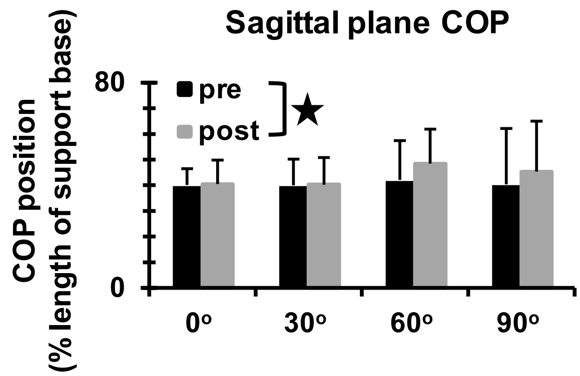


Fig. 3.11 COP position in the sagittal plane averaged across participants. Data are shown for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), before and after the application of the fatigue protocol. The black bars represent pre-fatigue data and the grey bars represent post-fatigue data. A value of 0 represents the COP being positioned at the bottom of the heels, whereas a value of 100 represents the COP being positioned at the tip of the toes. ★ : $p<0.05$. Note that there was also a significant knee position*fatigue interaction.

3.4.4 EMG

Fig. 3.12 displays the normalized RMS EMG activity, before and after fatigue, for all the recorded muscles, for the four different knee angle positions. Note that the averaged RMS values for all pre-fatigue and post-fatigue positions were normalized with the RMS value averaged across the three pre-fatigue trials at 90˚ of knee flexion for each muscle in each participant. Statistical analysis revealed that side (right/left) did not have a significant effect on any of the muscles [RF: $F(1,14)=1.3746$, $p=0.2605$; VM: $F(1,14)=1.8368$, $p=0.1967$; TA: $F(1,14)=2.7617$, $p=0.1187$; BF: $F(1,14)=0.0111$, $p=0.9174$; ST: $F(1,14)=0.7950$, $p=0.3876$; GM: $F(1,14)=2.4003$, $p=0.1436$, GL: $F(1,14)=0.4473$, $p=0.5144$].

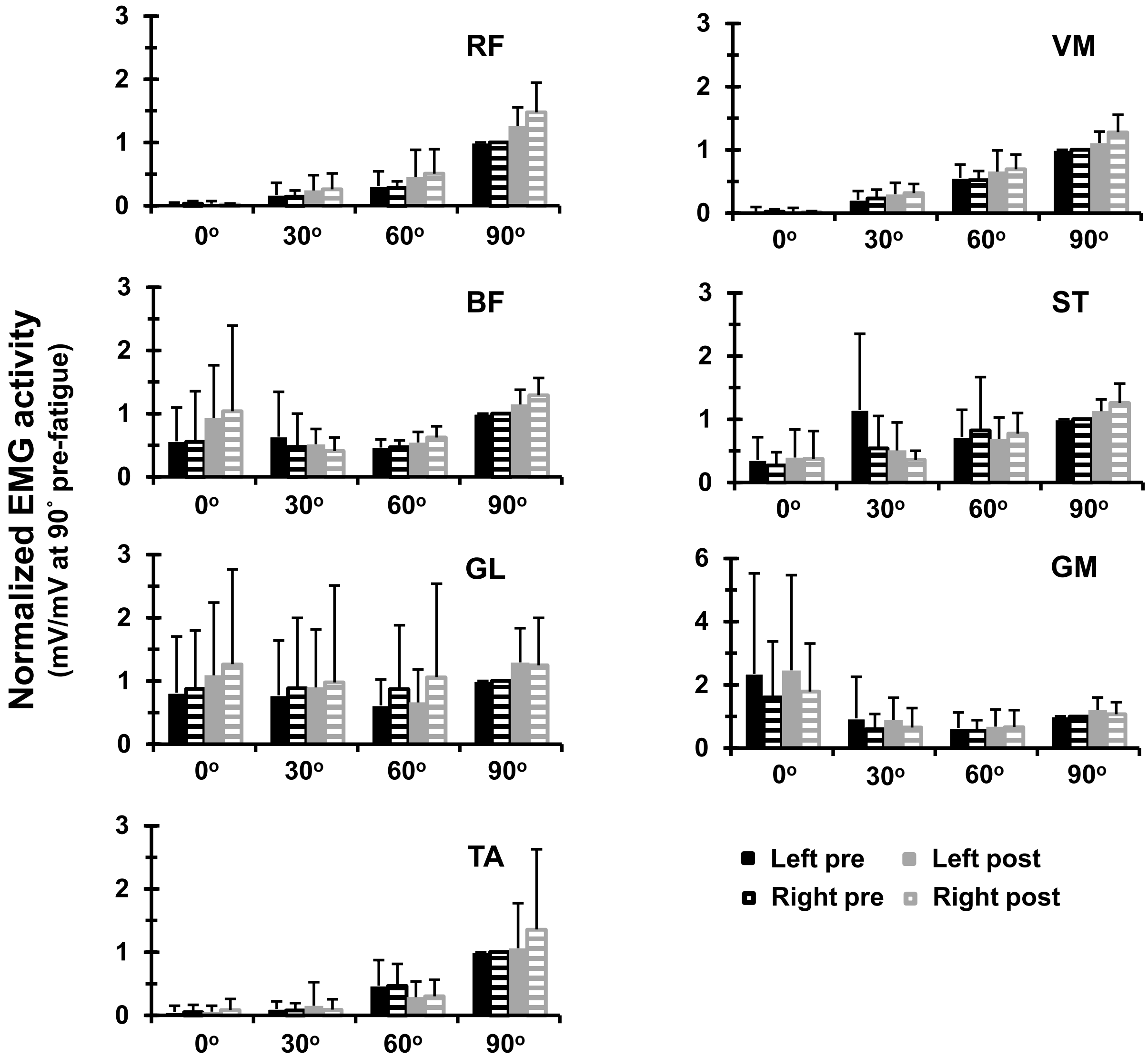


Fig. 3.12 RMS EMG activity for all the measured muscles averaged across participants. The values were normalized with the pre-fatigue value at 90˚. Data are shown for the four different knee angle positions (0˚, 30˚, 60˚, 90˚), before and after the application of the fatigue protocol. The solid bars represent the data from the left and the striped bars represent the data from the right leg. The black bars represent pre-fatigue data and grey bars represent post-fatigue data.

Table 3.2 shows significant main effects and interactions for all muscles. Except for GL, position had a significant effect on the studied muscles: (RF, $F(3,42)=216.0001$, $p<0.0001$; VM, $F(3,42)=253.6313$, $p<0.0001$; TA, $F(3,42)=62.0670$, $p<0.0001$; BF, $F(3,42)=12.6761$, $p<0.0001$; ST, $F(3,42)=23.5753$, $p<0.0001$; GM, $F(3,42)=5.8078$, $p=0.0002$; GL, $F(3,42)=0.8871$, $p=0.4556$). For RF, VM, ST, and TA, activity increased as knee angle increased (see Fig. 3.12).

For GM muscle activity was higher for 0^o whereas for BF muscle activity was higher for 90^o as compared to all other positions.

Table 3.2. Muscle activity statistical analysis

MUSCLES	SIDE (R/L)	FATIGUE (pre/post)	KNEE POSITION (0˚, 30˚, 60˚, 90˚)	FATIGUE*KNEE POSITION
RF		F(1,14)=14.14, p=0.0021	F(3,42)=216.0, p<0.0001	F(3,42)=11.8692, p<0.0001
VM		F(1,14)=18.95, p=0.0006	F(3,42)=253.63, p<0.0001	F(3,42)=8.7046, p=0.0001
ST			F(3,42)=23.57, p<0.0001	F(3,42)=5.5872, p=0.0025
BF		F(1,14)=22.55, p=0.0003	F(3,42)=12.67, p<0.0001	
TA			F(3,42)=62.06, p<0.0001	
GL		F(1,14)=7.6, p=0.0154		
GM			F(3,42)=5.8078, p=0.0002	

Statistical analysis also showed that fatigue provoked a significant increase in the activity of RF, VM, BF, and GL, but did not affect the amplitude of activity in other muscles [RF: $F(1,14)=14.1445$, $p=0.0021$; VM: $F(1,14)=18.9545$, $p=0.0006$; BF: $F(1,14)=22.5568$, $p=0.0003$; GL: $F(1,14)=7.6061$, $p=0.0154$; TA: $F(1,14)=0.05825$, $p=0.8127$; ST: $F(1,14)=0.3789$, $p=0.5480$; GM: $F(1,14)=1.9591$, $p=0.1833$].

There was also a significant position∗fatigue interaction for RF, VM, and ST [RF, $F(3,42)=11.8692$, $p<0.0001$; VM, $F(3,42)=8.7046$, $p=0.0001$; ST, $F(3,42)=5.5872$, $p=0.0025$; TA, $F(3,42)=1.5875$, $p=0.2066$; BF, $F(3,42)=2.7987$, $p=0.0516$; GM, $F(3,42)=0.2745$, $p=0.8434$; GL, $F(3,42)=1.0841$, $p=0.3662$]. For RF and VM the increase in activity post-fatigue was more elevated for greater knee angles (see Fig. 3.12). For ST muscle activity was decreased post-fatigue at 30˚.

3.5 Discussion

From the Borg scale results it is indicated that muscular fatigue was perceived by the participants, both immediately after the fatigue protocol and at the end of the post-fatigue trials, as these were both significantly greater than the two pre-fatigue values (see Fig. 3.7). The decrease in perceived fatigue in recording T4 as compared to recording T3 may reflect the starting of the recovery process; however, the recording T4 measurement (post-fatigue, after the second set of squats), was significantly greater than pre-fatigue recordings (T1-2) and was rated as "strong" indicating that fatigue was still perceived at the end of data collection.

The fatigue procedure led to a modification of COP position and of weight distribution in the antero-posterior direction but not in the coronal plane. There were also differences in EMG activation for some of the muscles post-fatigue. However, squatting position only produced differences in EMG activity, but not in weight distribution.

3.5.1 Weight distribution and COP

Coronal plane

Fatigue did not affect weight distribution in the coronal plane, which was expected, since the fatigue procedure was targeted bilaterally. There was therefore no biomechanical reason to distribute weight in favor of one side or the other. Fatiguing only one side would probably have led to a redistribution of the weight on the non-fatigued side. Indeed, Vuillerme (Vuillerme, et al., 2009) reported greater COP displacements, after a unilateral fatigue protocol, under the non-fatigued leg during bilateral stance, suggesting that participants supported a greater proportion of the weight on the non-fatigued side. Although weight was not perfectly distributed between right and left sides, our results show only a small asymmetry with results for all positions and conditions (pre/post fatigue) below 2.2%. Our results are similar to those reported in the literature with difference in weight distribution ranging from 1% to 4% in squatting or standing with similarly large standard deviations (Alexander & LaPier, 1998; Chmielewski, et al., 2002; Sackley, Lincoln, & 1991; Summers, Morrison, & Cochrane, 1987; Tessem, Hagstrom, & Fallang, 2007).

Sagittal plane

In our study, squatting position did not have an effect on global weight distribution in the antero-posterior direction nor on the position of the COP in the sagittal plane. This is in disagreement with a study by Dionisio (Dionisio, Almeida, Duarte, & Hirata, 2008), who reported the COP to stabilize at the tip of the toes at the end of dynamic squats. Dionisio observed a more anterior position of the hips, knees, and ankles at the end of the squat which probably led to the more anterior position of the COP. In our study, the flexibility in positioning the trunk may have led to the participants leaning forward, bringing the shoulders forward and the hips backward, with the COP being kept in the same position.

In our study, fatigue did not affect global weight distribution in the antero-posterior direction meaning that on average the same percentage of weight was borne under the anterior and posterior parts of the feet. However, we observed the COP to be positioned more anteriorly after the fatigue procedure. Our observation is in agreement with Madigan's findings that COP is positioned more anteriorly during quiet, bipedal stance, after fatiguing the lumbar extensors (Madigan, et al., 2006). According to the authors, this anterior shift of the COP causes an increase in plantarflexor muscle activity thereby increasing ankle stiffness, a necessary factor for controlling sway. The gastrocnemii also play a role in squats, although to a lesser degree than the quadriceps (Isear, Erickson, & Worrell, 1997). With an anterior shift of the COP post-fatigue, there would also be an increased load on the gastrocnemii, which may be reflected in the changes observed in our study (see below). In addition, it is also possible that this anterior shift of the COP might serve to biomechanically relieve the quadriceps by shifting the moment arm closer to the axis of rotation.

Although percent distribution on the posterior and anterior parts of the feet was not affected by fatigue, the amplitude of the posterior peak decreased whereas the amplitude of the anterior peak increased after the fatigue procedure. Therefore, although the same percentage of weight was borne under both the anterior and posterior parts of the feet before and after fatigue, these changes in amplitude indicate a more concentrated application of weight distribution anteriorly, and a more diffuse application posteriorly, after fatigue; the central area of the anterior part of the feet bears more pressure after fatigue, relieving the more peripheral areas, while the opposite is observed in the posterior part of the foot.

3.5.2 EMG

We observed no difference in activation of muscles from the left and right sides of the body. This was to be expected as there is no reason for a difference in activation in able-bodied individuals since both sides are equally capable of performing actions.

Activation of knee extensors and ankle dorsiflexors is required during squats in order to fight gravity and prevent falling backwards. Indeed, as reported by Dionisio (Dionisio, et al., 2008), we observed an increase in the activity of RF, VM and TA during squats. While Isear (Isear, et al., 1997) also found an increase in quadriceps activity with increasing knee angle, they recorded data during the descending, hold, and ascending phases of dynamic squats. It is therefore difficult to compare their results to ours, since there was a deceleration/acceleration factor which was not present in our study. Indeed, Gryzlo et al. (Gryzlo, Patek, Pink, & Perry, 1994) reported more activity for VM and VL during the ascending than during the descending phase of a squat although the participants were going through the same knee angle positions (90˚-0˚ vs. 0˚-90˚). The increased activity of the quadriceps in larger knee flexion angles can be explained by the fact that the quadriceps must generate more torque (and more force) as knee flexion increases to control the increasing moment arm (and torque) of the superimposed body weight at the knee joint (Levangie, 2011).

In our study, the knee angle position also led to a decrease in GM activation. This result is also not surprising since as knee angle increases during squats, plantarflexor activation is no longer required to prevent a forward fall.

As the load on the quadriceps is substantial during squats, the observed increase in the activity of the fatigued RF and VM might be expected as it is known that fatigued muscles require increased activation to generate the same amount of force (Izquierdo, et al., 2011), even though the anterior displacement of the COP squats after fatigue, might decrease the load on the knee extensors. The activity in GL may have increased because of the more anterior position after fatigue, which would increase the requirement for plantarflexor torque.

With the more anterior COP position adopted after fatigue, we would also expect the activity of the TA to be decreased since less dorsiflexor activation is required in this position. However, our data shows no effect of fatigue on TA activity. Although our fatigue protocol targeted mainly extensors, TA may have been active during the fatigue procedure, working in concert with the

GM and GL in order to stabilize the ankle. More TA activation may therefore have been required post-fatigue even to produce less force in a more anterior position.

The differential effect of fatigue and knee of position on the GM and GL muscles may also be due to physiological differences. Although it has been shown that these two muscles are similar with respect to fast and slow twitch fiber composition, they have other architectural differences (Edgerton, Smith, & Simpson, 1975; Johnson, Polgar, Weightman, & Appleton, 1973). GL has longer fascicle length but GM has larger fascicle angle and also can pack more muscle fibers within a certain volume, which gives GM the ability for greater force production (Kawakami, Ichinose, & Fukunaga, 1998). Since GM is a stronger muscle, it may have been less affected by the fatigue procedure and therefore produced a similar amount of activity pre and post-fatigue.

3.6 Conclusions

Our results show that fatigue did not affect weight distribution whether in the coronal plane or in the sagittal plane. However, although fatigue did not affect weight distribution in the sagittal plane, it did modify COP position. Indeed, the COP was positioned more anteriorly after the fatigue procedure. Moreover, both position and fatigue had an effect on EMG activity of some muscles. These changes may lead to increased risk of injury in athletes, when they become fatigued during practices or games, especially in deeper squat positions.

Chapter 4 – Conclusions

Muscular fatigue is a natural consequence of prolonged exercise which can lead to decreased force production (Rahnama, et al., 2003). It has also been proven to increase muscle reflex time and amplitude, as well as decrease proprioception, probably as a consequence of affecting several neural receptors (Lattanzio, et al., 1997; Melnyk & Gollhofer, 2007; Miura, et al., 2004; Mohammadi & Roozdar, 2010; Moore, et al., 2002; Ribeiro, et al., 2007; Rozzi, et al., 1999a; Skinner, et al., 1986; Wojtys, et al., 1996). Furthermore, from studies on fatigue it has been concluded that fatigue reduces balance during both unilateral and bilateral stance (Corbeil, et al., 2003; Dickin & Doan, 2008; Mello, et al., 2007; Springer & Pincivero, 2009; Yaggie & McGregor, 2002). Although balance is decreased during quiet standing after fatigue, we do not know how it is affected in other positions, such as during squats.

Our goal was to define and compare, during static squats, the weight distribution, the COP position and the activation of lower-limb muscles in a healthy population before and after a fatigue protocol. During squats, joints of the lower limbs are flexed and the extensors are under greater stress, fighting gravity to maintain the squatting position and also to maintain the center of gravity within the base of support. The torques produced about the hip, knee and ankle joints by the weight of the upper body during squats can be seen in Fig.4.1. According to Winter (Winter, 2009), 15 muscles are responsible for the sagittal plane torques at the hip, knee and ankle joints. During weight bearing situations, like squats, all three joint torques control the knee angle. This is why we believe that studying squats would be useful and would help draw conclusions about the biomechanical alterations that fatigue causes to the body during activities. Moreover, studying squats is interesting because they are commonly used for the rehabilitation procedure of different injuries. Squat is a closed-kinetic exercise which from a biomechanical point of view is suggested to be safer and to produce less stress and forces on joints which offers reduced risk to the recovering structures when compared to open-kinetic exercises (Sousa et al., 2007). Furthermore, the squat movement is commonly found in the routine of movements during different sports, such as in soccer when the goaltender is usually in a semi-squat position or in football and rugby when the front-line defence men are in a semi-squat position while they are waiting to tackle the players of the opposite team.

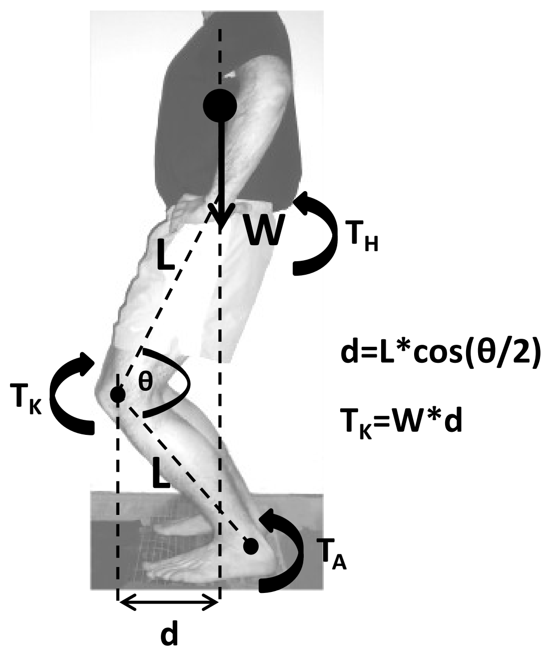


Fig. 4.1 Torques produced by muscles about the hip, knee, and ankle joints to counteract torques created by gravity. T_H is the sum of torques produced by the iliopsoas, gluteus maximus, semitendinosus, semimembranosus, biceps femoris, sartorius and rectus femoris. T_K is the sum of torques produced by the semitendinosus, semimembranosus, biceps femoris, sartorius, rectus femoris, vastus medialis, vastus lateralis, vastus intermedius and gastrocnemius. T_A is the sum of torques produced by the gastrocnemius, soleus, tibialis posterior, peronei and tibialis anterior (Winter, 2009). More information is provided on T_K, to support our discussion on the effect of knee position and fatigue on the quadriceps muscle. d is the distance between the knee center of rotation and the center of mass; W is body weight. Note that for simplicity it is assumed that the thigh and shank are the same length (L).

Our results showed that weight distribution was not affected by the fatigue procedure. However, the COP was positioned more anteriorly post-fatigue. The fact that fatigue did not affect weight distribution in the coronal plane was expected since the fatigue procedure targeted both sides. There was therefore no biomechanical reason to distribute weight in favor of one side or the other. We believe that the reason for the COP to be positioned more anteriorly after fatigue is for the individuals to biomechanically relieve their fatigued quadriceps. It can be seen in Fig. 4.2, that when the body leans more anteriorly (either leaning the trunk forward or by bringing the hips more forward), the torque produced about the knee by the weight of the body is smaller, meaning that the quadriceps need to be less active in order to counteract the torque produced by gravity.

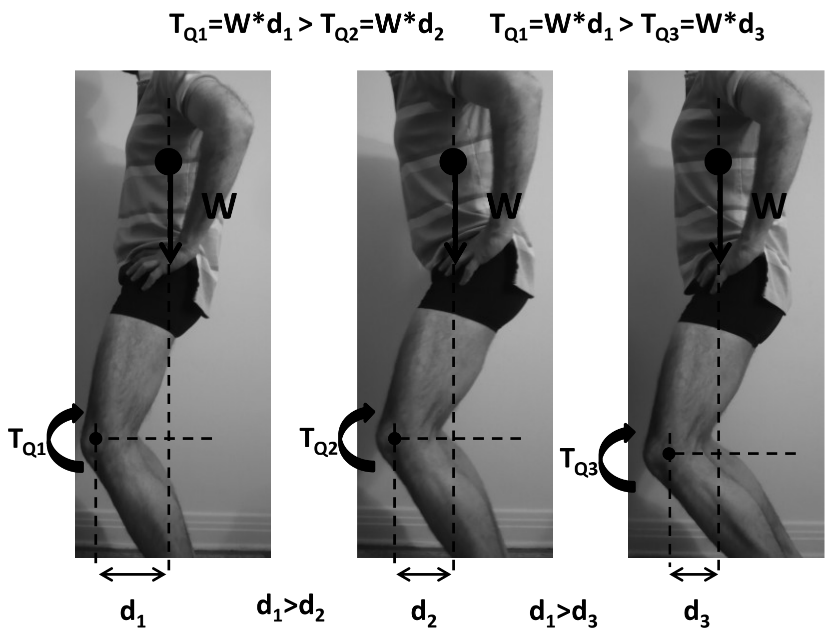


Fig. 4.2 Torques produced from body weight on the knee joint, when the body stays straight (left panel), when the trunk leans more anteriorly (middle panel) or when the hips are brought forward (right panel).

The suggestion that muscle load is decreased because of the COP being more forward may at first seem contradictory to our results that RF and VM were more active after the fatigue procedure. However, we suggest that the increase in RF and VM activity in our study was due to the fatigued muscles requiring more activation even to produce less torque. Note that one of the limitations of our study is that we did not record kinematics. Not knowing the position of the trunk, we do not know whether the more anterior position of the COP post-fatigue is due to the trunk leaning forward or to moving the hips more in front.

Our results also show an effect of position on muscle activation. More precisely, we found that there was an increase in the activity of RF, VM and TA as knee angle increased. As can be seen in Fig. 4.3, the flexor torque produced about the knee by the weight of the upper body during squats increases with knee flexion. Greater activation of knee extensors is thus required during squats in order to fight gravity and prevent falling. TA activation could be required in deeper squat positions to produce dorsiflexor torque and prevent falling backward although our data did not show backward displacement of the COP in greater knee angles.

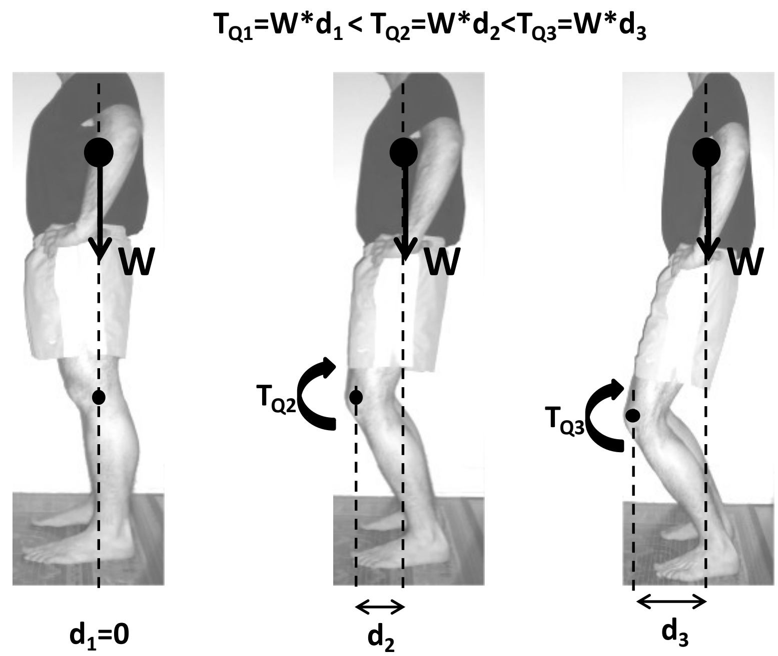


Fig. 4.3 Torque about the knee in different squatting positions. In deeper squat position, the torque produced by gravity on the knee joint is larger.

A fatigue protocol, like the one we used, could be expanded to injured individuals, where the differences in weight distribution and muscle activation between healthy and injured participants could be exacerbated due to muscular fatigue. For example, looking through the literature, we see that the stability of a person in unilateral stance on the involved limb after an ACL injury is significantly affected and decreased (Ageberg, Roberts, Holmstrom, & Friden, 2005; Gauffin, Pettersson, Tegner, & Tropp, 1990; Lysholm, Ledin, Odkvist, & Good, 1998; Mizuta, Shiraishi, Kubota, Kai, & Takagi, 1992; Okuda, et al., 2005) while dynamic stability is unaffected during bilateral stance (Lysholm, et al., 1998; Okuda, et al., 2005). Thus it is possible that the non-injured lower extremity may compensate for the injured leg. During bilateral static squats, it has been reported that weight distribution is not different in ACL-deficient individuals (Chmielewski, et al., 2002) nor are vertical ground reaction forces different on the two limbs of individuals who had undergone ACL reconstruction (Salem, Salinas, & Harding, 2003).

However, the latter authors observed that the moments produced at the lower-limb joints were different when comparing the two limbs. They concluded that there were two strategies used: the non-involved limb distributed the muscular effort between the hip and the knee, while the involved limb increased the effort at the hip and decreased it at the knee. They concluded that consistent use of the altered strategy might limit recovery and induce strength deficits. As well, in Chmielewski's study there was a correlation between weight distribution and quadriceps strength in ACL-deficient individuals. Taken together, we believe that muscular fatigue of the lower-limb extensors could affect weight distribution in ACL-deficient patients and induce strategies for maintaining stability that might be counterproductive to full rehabilitation.

Evaluation of how weight distribution is affected after a fatigue protocol might be used to mimic the fatigue that the athletes might feel during sports situations such as during a game. Thus it may be possible to use this technique as a screening procedure before deciding if any athlete, after injury rehabilitation, is ready to return to training and games. Also, by measuring the activity of specific leg muscles, it may be possible to see how muscle activity is different after fatigue so that possible compensations used by injured individuals to control stability may be described. Moreover, since squats are a typical closed-kinetic exercise which is used broadly in the rehabilitation procedure of ligamentous injuries in the knee joint (i.e. ACL rupture or ACL reconstruction), it would be interesting to know how weight is distributed and how muscles are activated during this exercise in order to provide better guidelines for health professionals.

In our study, we observed no difference in activation of muscles from the left and right sides of the body. This was to be expected as there is no reason for a difference in activation in able-bodied individuals since both sides are equally capable of performing actions. Evaluating injured populations may lead to the observation of differences between the right and left sides as the sound limb may be compensating for the injured limb. Having measurements during bilateral stance or squats, where possible injuries can be "masked" due to compensation by the contralateral leg, can give a better idea of what might be happening in unilateral injuries. In future experiments, it may be interesting to study dynamic movements such as walking, running or jumping which may be more relevant and more representative of sports situations. However, in the future, very useful observations could also be extracted from studies focusing on single-legged stance and movements.

As with any study, there were limitations with this project. The major limitation of this study is the fact that our participants were only relatively young, physically active, men. This limits our results to a specific age group, with a specific physical activity level and of course refers only to one gender. In the future a similar study would be more complete if women were included and also less active people from different age groups. A major limitation regarding the methods of this project was that we did not control for the angle and the distance between the feet as well as for trunk orientation while the participants were executing the squats. We opted not to control for the position of the feet and trunk so that the posture of the participants would be more natural and the experiment would simulate real life conditions. However, the distance between the heels was only slightly affected by fatigue (pre-fatigue mean distance: 28.8cm, post-fatigue mean distance: 28cm), whereas the angle was not. Furthermore, another limitation was that participants executed the different squat positions in the same order (0˚, 30˚, 60˚, 90˚). This lack of randomization in the order of the positions could potentially have affected our results. Another limitation was that the fatigue protocol targeted only the muscles that participate in the performance of squats (mainly extensors). In the future, it may be interesting to study fatigue of other muscles. For example, since hamstrings restrict anterior tibial translation, it would be interesting to study the effect of fatiguing this muscle group especially in ACL-deficient individuals. Moreover, it would have been very useful to have kinematic data for the trunk position, because with these data we could show whether the forward shift of the COP position is due to leaning of the trunk or to a more forward position of the hips. It would also have been interesting to record EMG activity of the gluteus maximus and the soleus muscles. These muscles although they do not cross the knee joint are capable of assisting with knee extension. It is well known that during weight-bearing exercises, the soleus contraction can assist with knee extension by pulling the tibia posteriorly and the gluteus maximus contraction also helps with the knee extension by producing a posterior shear of the femur on the tibia (Levangie, 2011). Finally, it may be interesting to study other movements during dynamic unilateral stance, usually present during athletic performance, like walking, jumping, cutting or running.

Overall, this study can serve as a basis for future research on the effect of fatigue on weight distribution and muscle activity. It could be expanded to the evaluation of different musculoskeletal injuries that may affect weight distribution. For example unilateral injuries of ligaments, tendons and muscles can be "masked" during bilateral stance and activities and may provide the therapists with post-injury outcomes influenced by weight distribution compensation. For the injured individual, it might be expected that this mechanism would be exacerbated due to fatigue, and differences in weight distribution and muscle activation might also be observed. Monitoring these factors during post-injury training could serve as a useful evaluation tool for return to sports for injured populations. Studying injured individuals after a fatigue protocol, which resembles real life conditions during games, it may be possible to decide if a person, after an injury, is ready to return to sports or is still at risk and should avoid returning to sports. Moreover, since valgus-varus moments are present during squats, this study can be expanded to females who because of anatomical differences have a larger valgus angle than men. It would be interesting to study squats in women and see what the differences are with men. These differences could help explain why women are more prone to some injuries such as ACL ruptures.

References

Adlerton, A. K., & Moritz, U. (1996). Does calf-muscle fatigue affect standing balance? *Scand J Med Sci Sports, 6*(4), 211-215.

Ageberg, E., Roberts, D., Holmstrom, E., & Friden, T. (2005). Balance in single-limb stance in patients with anterior cruciate ligament injury: relation to knee laxity, proprioception, muscle strength, and subjective function. *Am J Sports Med, 33*(10), 1527-1535.

Alexander, K. M., & LaPier, T. L. (1998). Differences in static balance and weight distribution between normal subjects and subjects with chronic unilateral low back pain. *J Orthop Sports Phys Ther, 28*(6), 378-383.

Bigland-Ritchie, B. (1981). EMG/force relations and fatigue of human voluntary contractions. *Exerc Sport Sci Rev, 9*, 75-117.

Borg, G. (1970). Perceived exertion as an indicator of somatic stress. *Scand J Rehabil Med, 2*(2), 92-98.

Carling, C., Gall, F. L., & Reilly, T. P. (2010). Effects of physical efforts on injury in elite soccer. *Int J Sports Med, 31*(3), 180-185.

Chappell, J. D., Herman, D. C., Knight, B. S., Kirkendall, D. T., Garrett, W. E., & Yu, B. (2005). Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *Am J Sports Med, 33*(7), 1022-1029.

Chmielewski, T. L., Wilk, K. E., & Snyder-Mackler, L. (2002). Changes in weight-bearing following injury or surgical reconstruction of the ACL: relationship to quadriceps strength and function. *Gait Posture, 16*(1), 87-95.

Colby, S. M., Hintermeister, R. A., Torry, M. R., & Steadman, J. R. (1999). Lower limb stability with ACL impairment. *J Orthop Sports Phys Ther, 29*(8), 444-451; discussion 452-444.

Corbeil, P., Blouin, J. S., Begin, F., Nougier, V., & Teasdale, N. (2003). Perturbation of the postural control system induced by muscular fatigue. *Gait Posture, 18*(2), 92-100.

Davidson, B. S., Madigan, M. L., & Nussbaum, M. A. (2004). Effects of lumbar extensor fatigue and fatigue rate on postural sway. *Eur J Appl Physiol, 93*(1-2), 183-189.

Dickin, D. C., & Doan, J. B. (2008). Postural stability in altered and unaltered sensory environments following fatiguing exercise of lower extremity joints. *Scand J Med Sci Sports, 18*(6), 765-772.

Dionisio, V. C., Almeida, G. L., Duarte, M., & Hirata, R. P. (2008). Kinematic, kinetic and EMG patterns during downward squatting. *J Electromyogr Kinesiol, 18*(1), 134-143.

Dobkin, B. H. (2008). Fatigue versus activity-dependent fatigability in patients with central or peripheral motor impairments. *Neurorehabil Neural Repair, 22*(2), 105-110.

Edgerton, V. R., Smith, J. L., & Simpson, D. R. (1975). Muscle fibre type populations of human leg muscles. *Histochem J, 7*(3), 259-266.

Eklund, G. (1972). General features of vibration-induced effects on balance. *Ups J Med Sci, 77*(2), 112-124.

Enoka, R. M., & Duchateau, J. (2008). Muscle fatigue: what, why and how it influences muscle function. *J Physiol, 586*(1), 11-23.

Gauffin, H., Pettersson, G., Tegner, Y., & Tropp, H. (1990). Function testing in patients with old rupture of the anterior cruciate ligament. *Int J Sports Med, 11*(1), 73-77.

Gosselin, G., Rassoulian, H., & Brown, I. (2004). Effects of neck extensor muscles fatigue on balance. *Clin Biomech (Bristol, Avon), 19*(5), 473-479.

Granata, K. P., & Gottipati, P. (2008). Fatigue influences the dynamic stability of the torso. *Ergonomics, 51*(8), 1258-1271.

Gribble, P. A., & Hertel, J. (2004). Effect of hip and ankle muscle fatigue on unipedal postural control. *J Electromyogr Kinesiol, 14*(6), 641-646.

Gryzlo, S. M., Patek, R. M., Pink, M., & Perry, J. (1994). Electromyographic analysis of knee rehabilitation exercises. *J Orthop Sports Phys Ther, 20*(1), 36-43.

Guyton, A., & Hall, J. (2006). *Textbook of medical physiology*. Philadelphia, PA, USA: Elsevier Inc.

Heuser, M., & Pincivero, D. (2010). The effects of stretching on knee flexor fatigue and perceived exertion. *J Sports Sci, 28*(2), 219-226.

Hiemstra, L. A., Lo, I. K., & Fowler, P. J. (2001). Effect of fatigue on knee proprioception: implications for dynamic stabilization. *J Orthop Sports Phys Ther, 31*(10), 598-605.

Hogervorst, T., & Brand, R. A. (1998). Mechanoreceptors in joint function. *J Bone Joint Surg Am, 80*(9), 1365-1378.

Isear, J. A., Jr., Erickson, J. C., & Worrell, T. W. (1997). EMG analysis of lower extremity muscle recruitment patterns during an unloaded squat. *Med Sci Sports Exerc, 29*(4), 532-539.

Izquierdo, M., Gonzalez-Izal, M., Navarro-Amezqueta, I., Calbet, J. A., Ibanez, J., Malanda, A., et al. (2011). Effects of strength training on muscle fatigue mapping from surface EMG and blood metabolites. *Med Sci Sports Exerc, 43*(2), 303-311.

Johnson, M. A., Polgar, J., Weightman, D., & Appleton, D. (1973). Data on the distribution of fibre types in thirty-six human muscles. An autopsy study. *J Neurol Sci, 18*(1), 111-129.

Kawakami, Y., Ichinose, Y., & Fukunaga, T. (1998). Architectural and functional features of human triceps surae muscles during contraction. *J Appl Physiol, 85*(2), 398-404.

Kellis, E., & Liassou, C. (2009). The effect of selective muscle fatigue on sagittal lower limb kinematics and muscle activity during level running. *J Orthop Sports Phys Ther, 39*(3), 210-220.

Kennedy, J. C., Alexander, I. J., & Hayes, K. C. (1982). Nerve supply of the human knee and its functional importance. *Am J Sports Med, 10*(6), 329-335.

Kent-Braun, J. A. (1999). Central and peripheral contributions to muscle fatigue in humans during sustained maximal effort. *Eur J Appl Physiol Occup Physiol, 80*(1), 57-63.

Lattanzio, P. J., Petrella, R. J., Sproule, J. R., & Fowler, P. J. (1997). Effects of fatigue on knee proprioception. *Clin J Sport Med, 7*(1), 22-27.

Levangie, P. K., Norkin, C.C. (Ed.). (2011). *Joint structure and function - a comprehensive analysis* (5th ed.). Philadelphia, PA: Davis Company.

Lysholm, M., Ledin, T., Odkvist, L. M., & Good, L. (1998). Postural control--a comparison between patients with chronic anterior cruciate ligament insufficiency and healthy individuals. *Scand J Med Sci Sports, 8*(6), 432-438.

Madigan, M. L., Davidson, B. S., & Nussbaum, M. A. (2006). Postural sway and joint kinematics during quiet standing are affected by lumbar extensor fatigue. *Hum Mov Sci, 25*(6), 788-799.

Mannion, A. F., & Dolan, P. (1996). Relationship between myoelectric and mechanical manifestations of fatigue in the quadriceps femoris muscle group. *Eur J Appl Physiol Occup Physiol, 74*(5), 411-419.

McLean, S. G., Fellin, R. E., Suedekum, N., Calabrese, G., Passerallo, A., & Joy, S. (2007). Impact of fatigue on gender-based high-risk landing strategies. *Med Sci Sports Exerc, 39*(3), 502-514.

Mello, R. G., Oliveira, L. F., & Nadal, J. (2007). Anticipation mechanism in body sway control and effect of muscle fatigue. *J Electromyogr Kinesiol, 17*(6), 739-746.

Melnyk, M., & Gollhofer, A. (2007). Submaximal fatigue of the hamstrings impairs specific reflex components and knee stability. *Knee Surg Sports Traumatol Arthrosc, 15*(5), 525-532.

Miura, K., Ishibashi, Y., Tsuda, E., Okamura, Y., Otsuka, H., & Toh, S. (2004). The effect of local and general fatigue on knee proprioception. *Arthroscopy, 20*(4), 414-418.

Mizuta, H., Shiraishi, M., Kubota, K., Kai, K., & Takagi, K. (1992). A stabilometric technique for evaluation of functional instability in anterior cruciate ligament-deficient knee. *Clin J Sport Med, 2*, 235-239.

Mohammadi, F., & Roozdar, A. (2010). Effects of fatigue due to contraction of evertor muscles on the ankle joint position sense in male soccer players. *Am J Sports Med, 38*(4), 824-828.

Moore, B. D., Drouin, J., Gansneder, B. M., & Shultz, S. J. (2002). The differential effects of fatigue on reflex response timing and amplitude in males and females. *J Electromyogr Kinesiol, 12*(5), 351-360.

Moran, K. A., Clarke, M., Reilly, F., Wallace, E. S., Brabazon, D., & Marshall, B. (2009). Does endurance fatigue increase the risk of injury when performing drop jumps? *J Strength Cond Res, 23*(5), 1448-1455.

Ochsendorf, D. T., Mattacola, C. G., & Arnold, B. L. (2000). Effect of orthotics on postural sway after fatigue of the plantar flexors and dorsiflexors. *J Athl Train, 35*(1), 26-30.

Okuda, K., Abe, N., Katayama, Y., Senda, M., Kuroda, T., & Inoue, H. (2005). Effect of vision on postural sway in anterior cruciate ligament injured knees. *J Orthop Sci, 10*(3), 277-283.

Orishimo, K. F., & Kremenic, I. J. (2006). Effect of fatigue on single-leg hop landing biomechanics. *J Appl Biomech, 22*(4), 245-254.

Ortiz, A., Olson, S. L., Etnyre, B., Trudelle-Jackson, E. E., Bartlett, W., & Venegas-Rios, H. L. (2010). Fatigue effects on knee joint stability during two jump tasks in women. *J Strength Cond Res, 24*(4), 1019-1027.

Padua, D. A., Arnold, B. L., Perrin, D. H., Gansneder, B. M., Carcia, C. R., & Granata, K. P. (2006). Fatigue, vertical leg stiffness, and stiffness control strategies in males and females. *J Athl Train, 41*(3), 294-304.

Pline, K. M., Madigan, M. L., & Nussbaum, M. A. (2006). Influence of fatigue time and level on increases in postural sway. *Ergonomics, 49*(15), 1639-1648.

Rahnama, N., Reilly, T., & Lees, A. (2002). Injury risk associated with playing actions during competitive soccer. *Br J Sports Med, 36*(5), 354-359.

Rahnama, N., Reilly, T., Lees, A., & Graham-Smith, P. (2003). Muscle fatigue induced by exercise simulating the work rate of competitive soccer. *J Sports Sci, 21*(11), 933-942.

Ribeiro, F., Mota, J., & Oliveira, J. (2007). Effect of exercise-induced fatigue on position sense of the knee in the elderly. *Eur J Appl Physiol, 99*(4), 379-385.

Rozzi, S. L., Lephart, S. M., & Fu, F. H. (1999a). Effects of Muscular Fatigue on Knee Joint Laxity and Neuromuscular Characteristics of Male and Female Athletes. *J Athl Train, 34*(2), 106-114.

Rozzi, S. L., Lephart, S. M., & Fu, F. H. (1999b). Effects of muscular fatigue on knee joint laxity and neuromuscular characteristics of male and female athletes. *J Athl Train, 34*(2), 106-114.

Sackley, C. M., Lincoln, N. B., & (1991). Weight distribution and postural sway in healthy adults. *Clin Rehabil, 5*, 181-186.
Salem, G. J., Salinas, R., & Harding, F. V. (2003). Bilateral kinematic and kinetic analysis of the squat exercise after anterior cruciate ligament reconstruction. *Arch Phys Med Rehabil, 84*(8), 1211-1216.
Schutte, M. J., Dabezies, E. J., Zimny, M. L., & Happel, L. T. (1987). Neural anatomy of the human anterior cruciate ligament. *J Bone Joint Surg Am, 69*(2), 243-247.
Skinner, H. B., Wyatt, M. P., Hodgdon, J. A., Conard, D. W., & Barrack, R. L. (1986). Effect of fatigue on joint position sense of the knee. *J Orthop Res, 4*(1), 112-118.
Small, K., McNaughton, L. R., Greig, M., Lohkamp, M., & Lovell, R. (2009). Soccer fatigue, sprinting and hamstring injury risk. *Int J Sports Med, 30*(8), 573-578.
Springer, B. K., & Pincivero, D. M. (2009). The effects of localized muscle and whole-body fatigue on single-leg balance between healthy men and women. *Gait Posture, 30*(1), 50-54.
Summers, G. D., Morrison, J. D., & Cochrane, G. M. (1987). Foot loading characteristics of amputees and normal subjects. *Prosthet Orthot Int, 11*(1), 33-39.
Suponitsky, Y., Verbitsky, O., Peled, E., & Mizrahi, J. (2008). Effect of selective fatiguing of the shank muscles on single-leg-standing sway. *J Electromyogr Kinesiol, 18*(4), 682-689.
Taylor, B. J., & Romer, L. M. (2008). Effect of expiratory muscle fatigue on exercise tolerance and locomotor muscle fatigue in healthy humans. *J Appl Physiol, 104*(5), 1442-1451.
Taylor, J. L., Todd, G., & Gandevia, S. C. (2006). Evidence for a supraspinal contribution to human muscle fatigue. *Clin Exp Pharmacol Physiol, 33*(4), 400-405.
Tessem, S., Hagstrom, N., & Fallang, B. (2007). Weight distribution in standing and sitting positions, and weight transfer during reaching tasks, in seated stroke subjects and healthy subjects. *Physiother Res Int, 12*(2), 82-94.
Troiano, A., Naddeo, F., Sosso, E., Camarota, G., Merletti, R., & Mesin, L. (2008). Assessment of force and fatigue in isometric contractions of the upper trapezius muscle by surface EMG signal and perceived exertion scale. *Gait Posture, 28*(2), 179-186.
Vuillerme, N., & Boisgontier, M. (2010). Changes in the relative contribution of each leg to the control of quiet two-legged stance following unilateral plantar-flexor muscles fatigue. *Eur J Appl Physiol, 110*(1), 207-213.
Vuillerme, N., Nougier, V., & Prieur, J. M. (2001). Can vision compensate for a lower limbs muscular fatigue for controlling posture in humans? *Neurosci Lett, 308*(2), 103-106.
Vuillerme, N., Sporbert, C., & Pinsault, N. (2009). Postural adaptation to unilateral hip muscle fatigue during human bipedal standing. *Gait Posture, 30*(1), 122-125.
Winter, D. A. (Ed.). (2009). *Biomechanics and motor control of human movement* (4th ed.). Hoboken, N.J.: Wiley.
Wojtys, E. M., Wylie, B. B., & Huston, L. J. (1996). The effects of muscle fatigue on neuromuscular function and anterior tibial translation in healthy knees. *Am J Sports Med, 24*(5), 615-621.
Yaggie, J. A., & McGregor, S. J. (2002). Effects of isokinetic ankle fatigue on the maintenance of balance and postural limits. *Arch Phys Med Rehabil, 83*(2), 224-228.

Appendix A

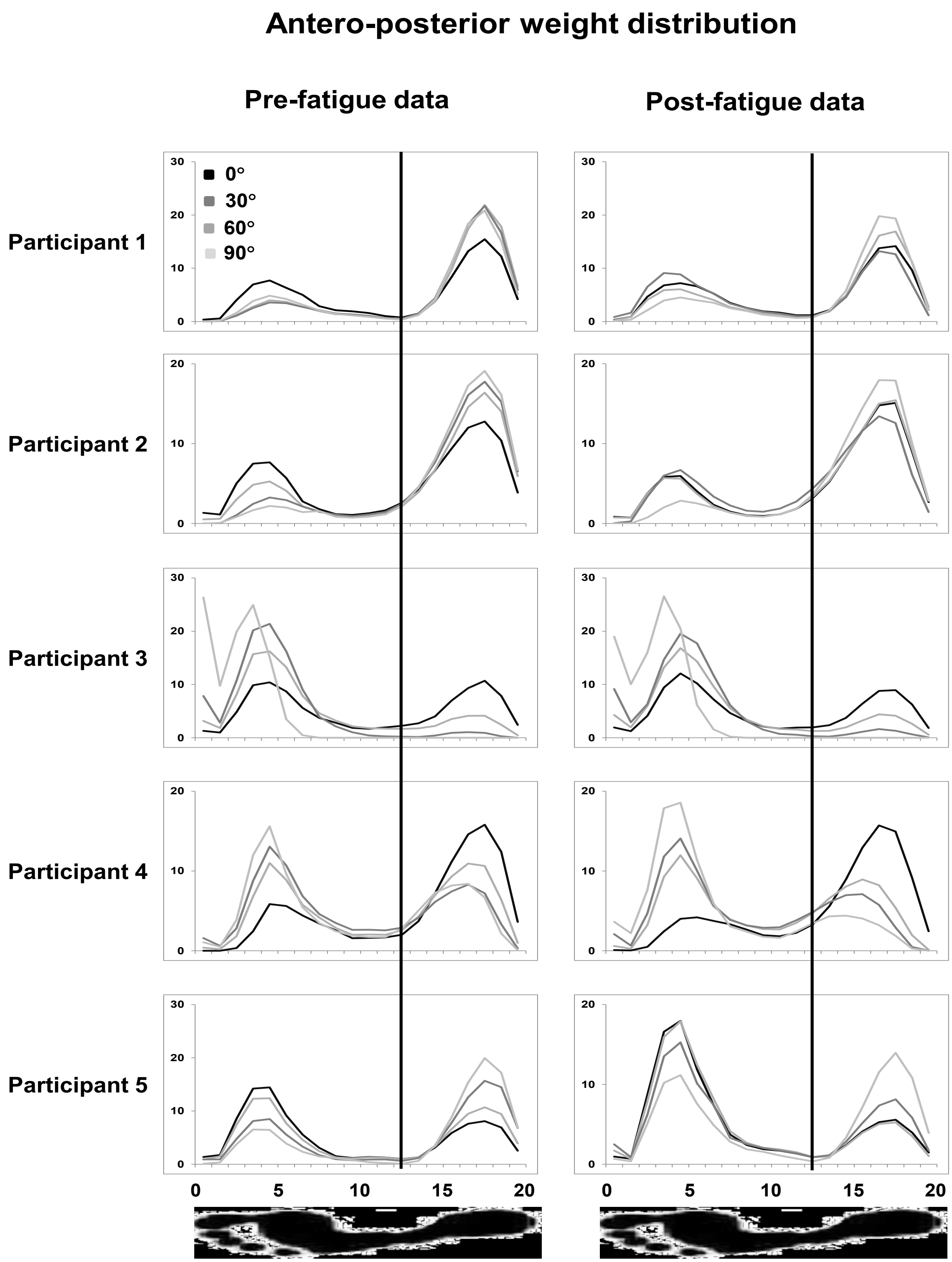


Figure A.1 Weight distribution data for each knee position in the sagittal plane, for every participant. The panels on the left represent data before the fatigue protocol, and the panels on the right represent data after the fatigue protocol. Feet were separated into 20 antero-posterior sections. Pressure under each section was normalized to body weight. The vertical lines show where it was chosen to split the feet into two parts: anterior (1-12) and posterior (13-20).

Antero-posterior weight distribution (cont'd)

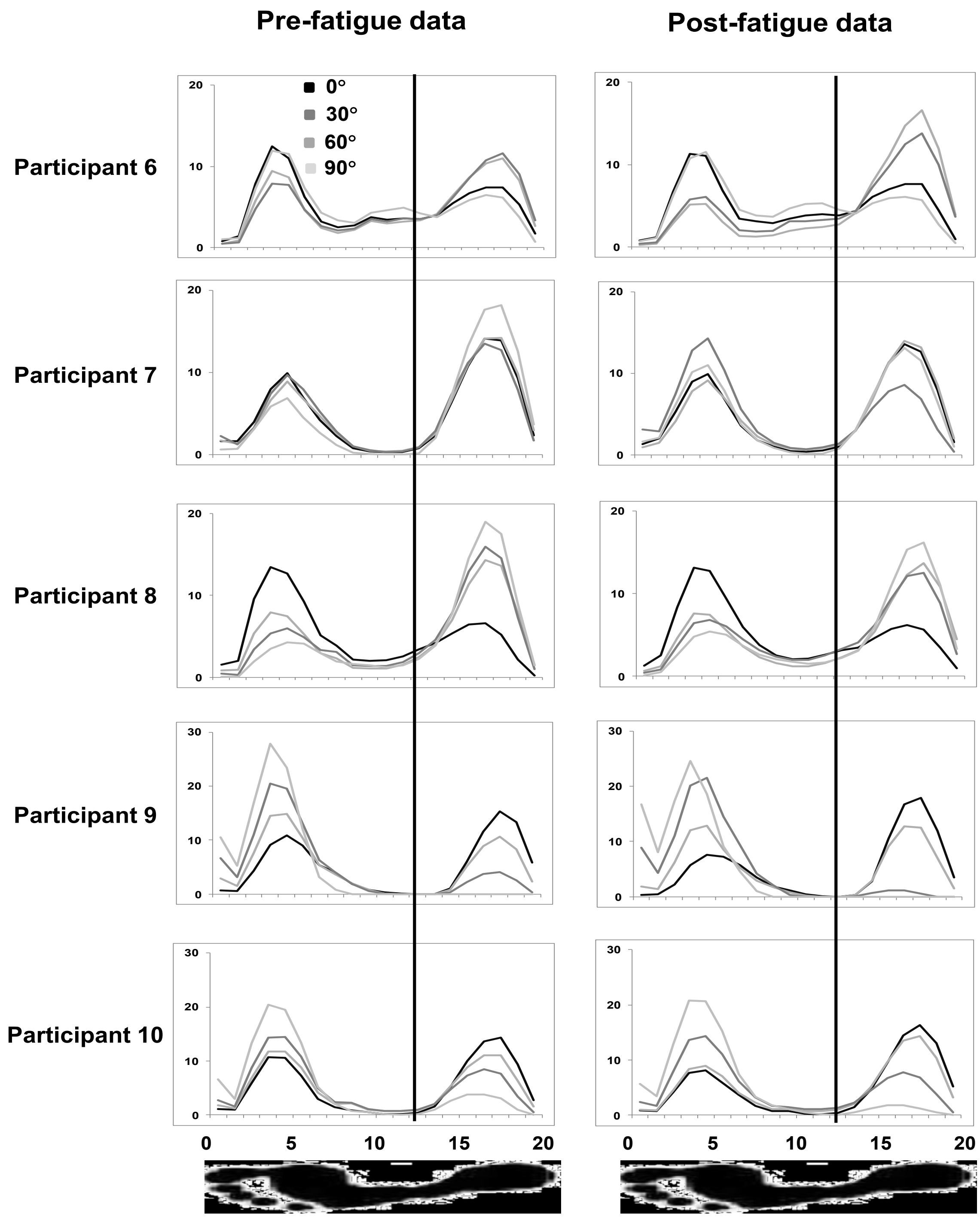


Figure A.1 cont'd

Antero-posterior weight distribution (cont'd)

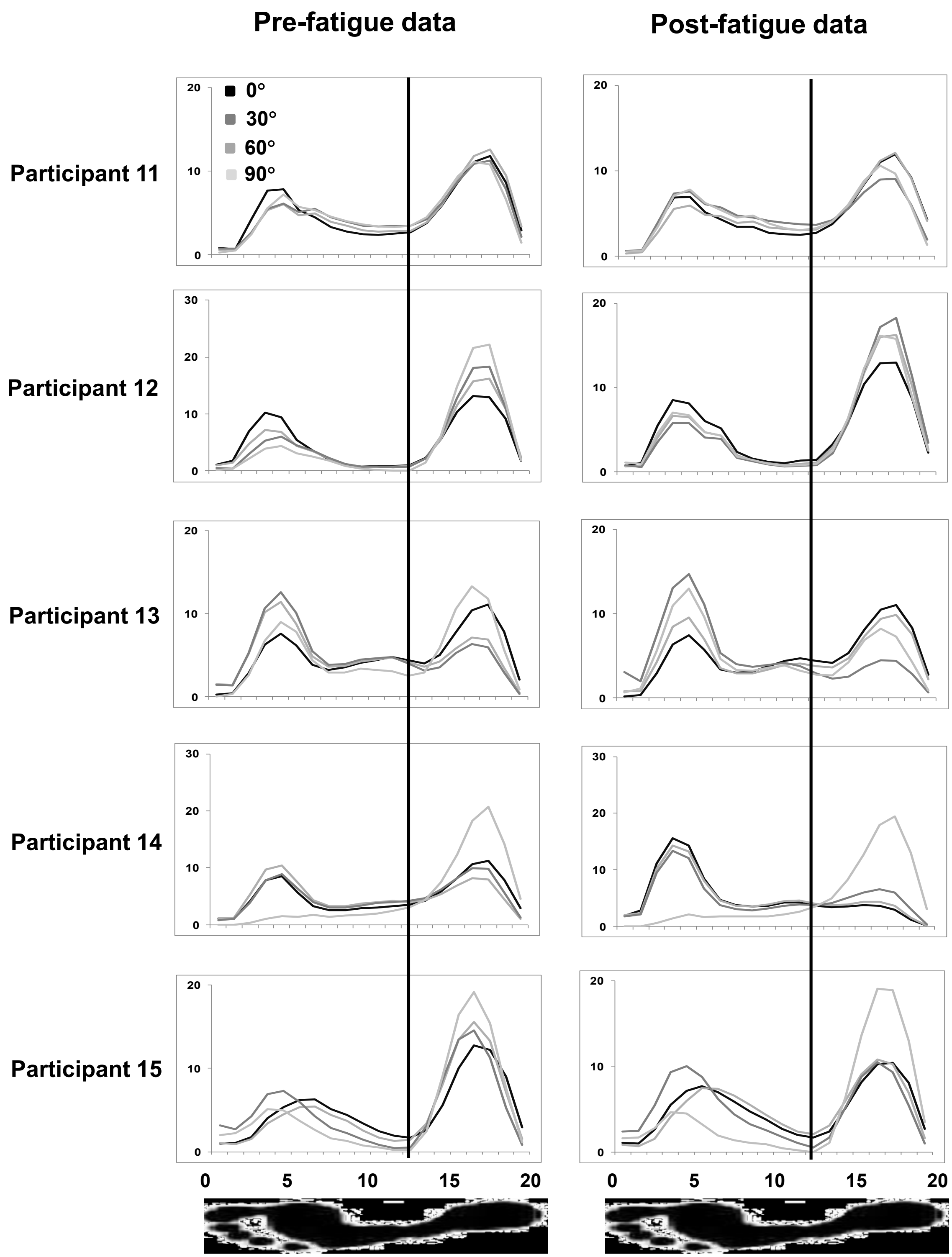


Figure A.1 cont'd

<u>Appendix B</u>

Normality graphs for non-dominant – dominant weight distribution

Pre-fatigue data

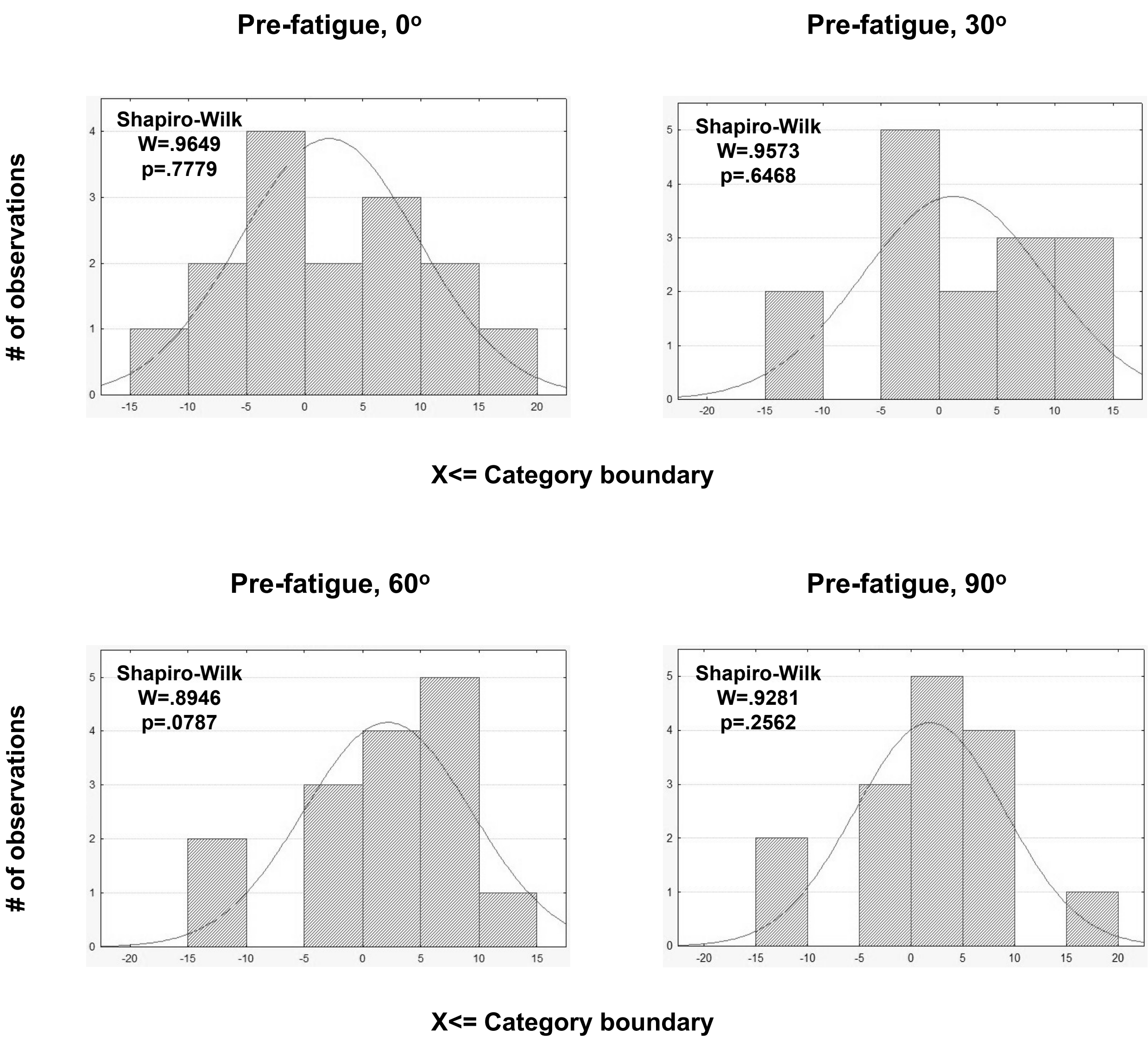


Figure B.1 Normality graphs for all participants for weight distribution in the coronal plane for each knee position, before the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for non-dominant – dominant weight distribution

Post-fatigue data

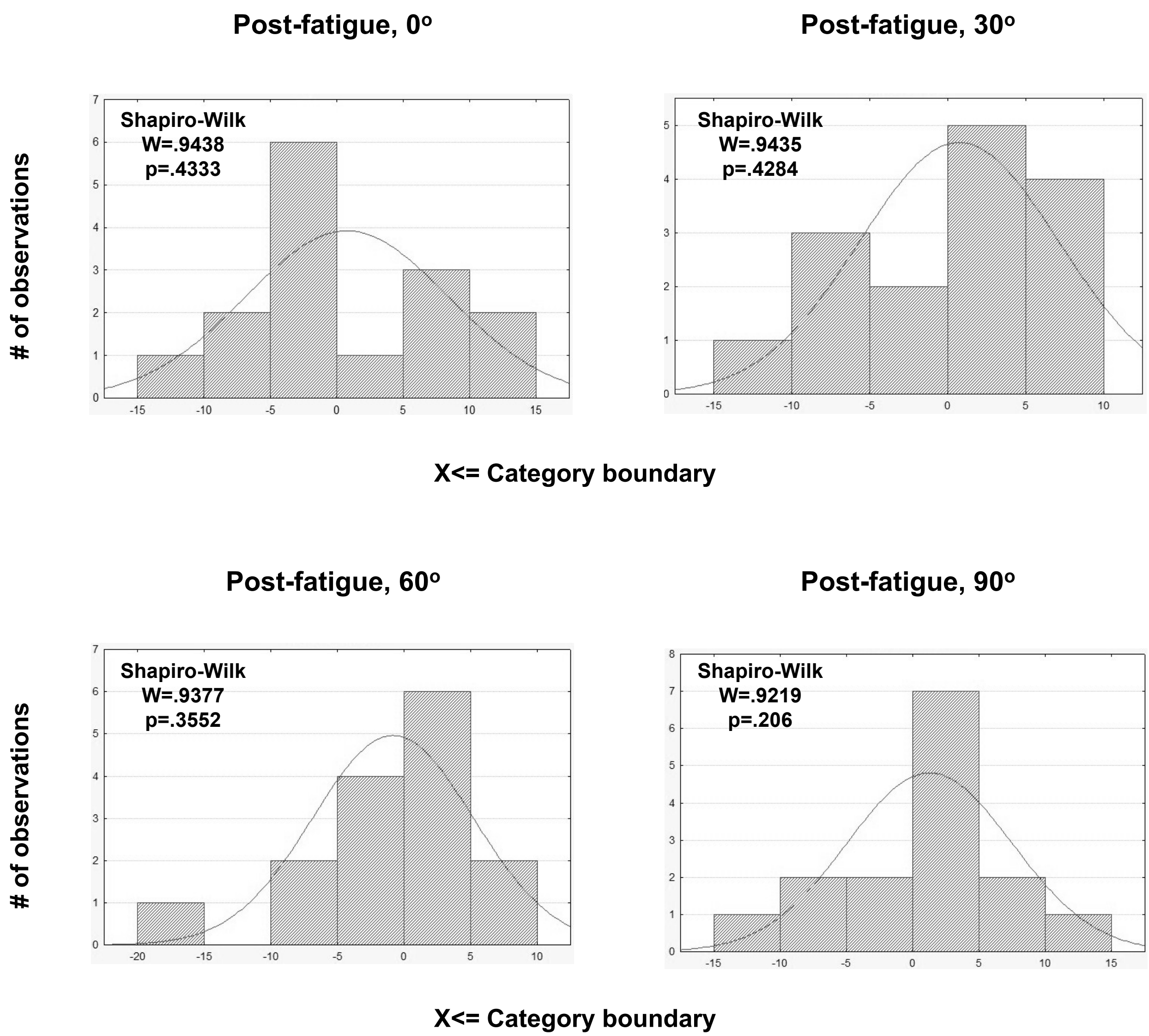


Figure B.2 Normality graphs for all participants for weight distribution in the coronal plane for each knee position, after the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for antero-posterior (global) weight distribution

Pre-fatigue data

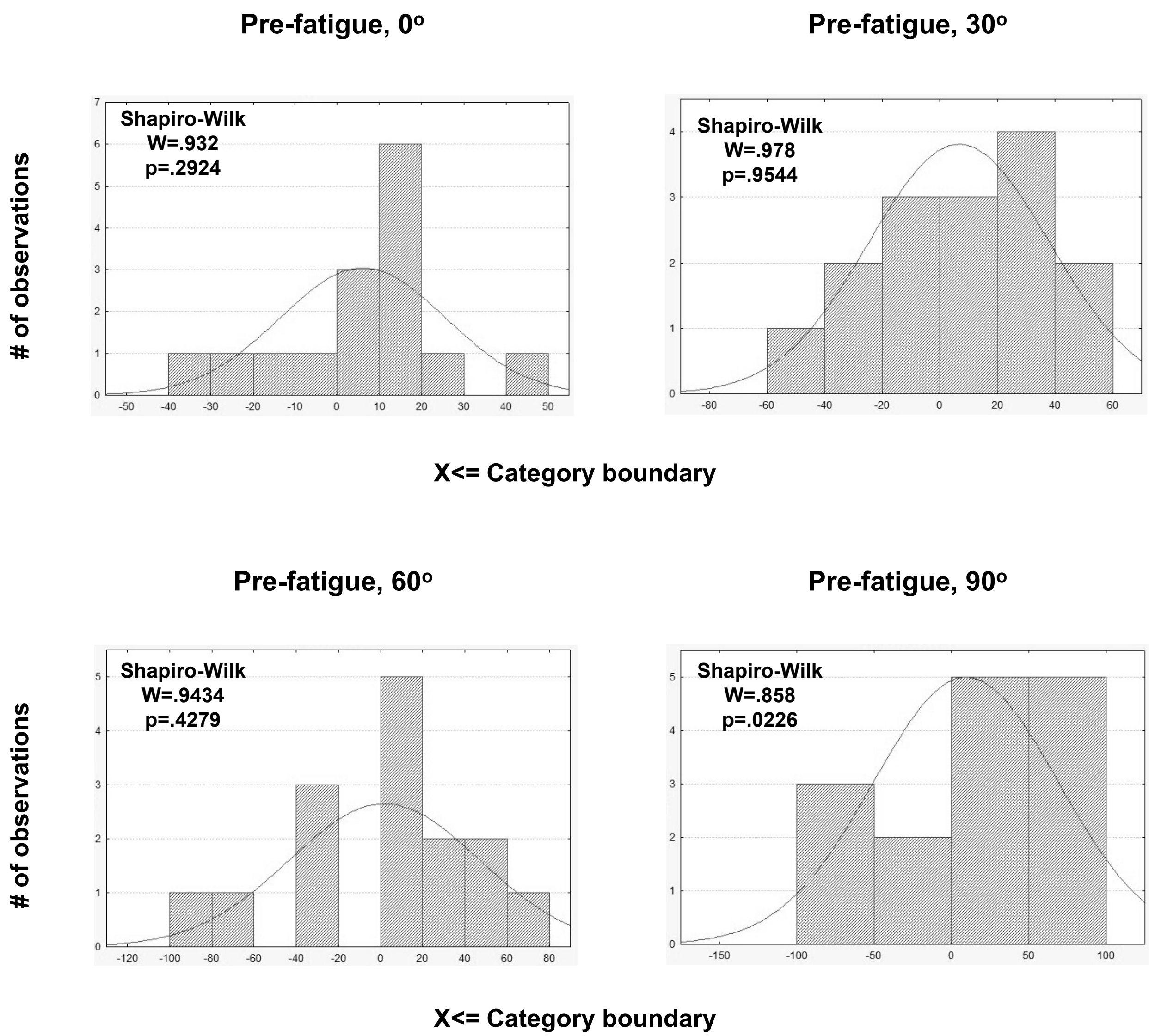


Figure B.3 Normality graphs for all participants for weight distribution in the sagittal plane for each knee position, before the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for antero-posterior (global) weight distribution

Post-fatigue data

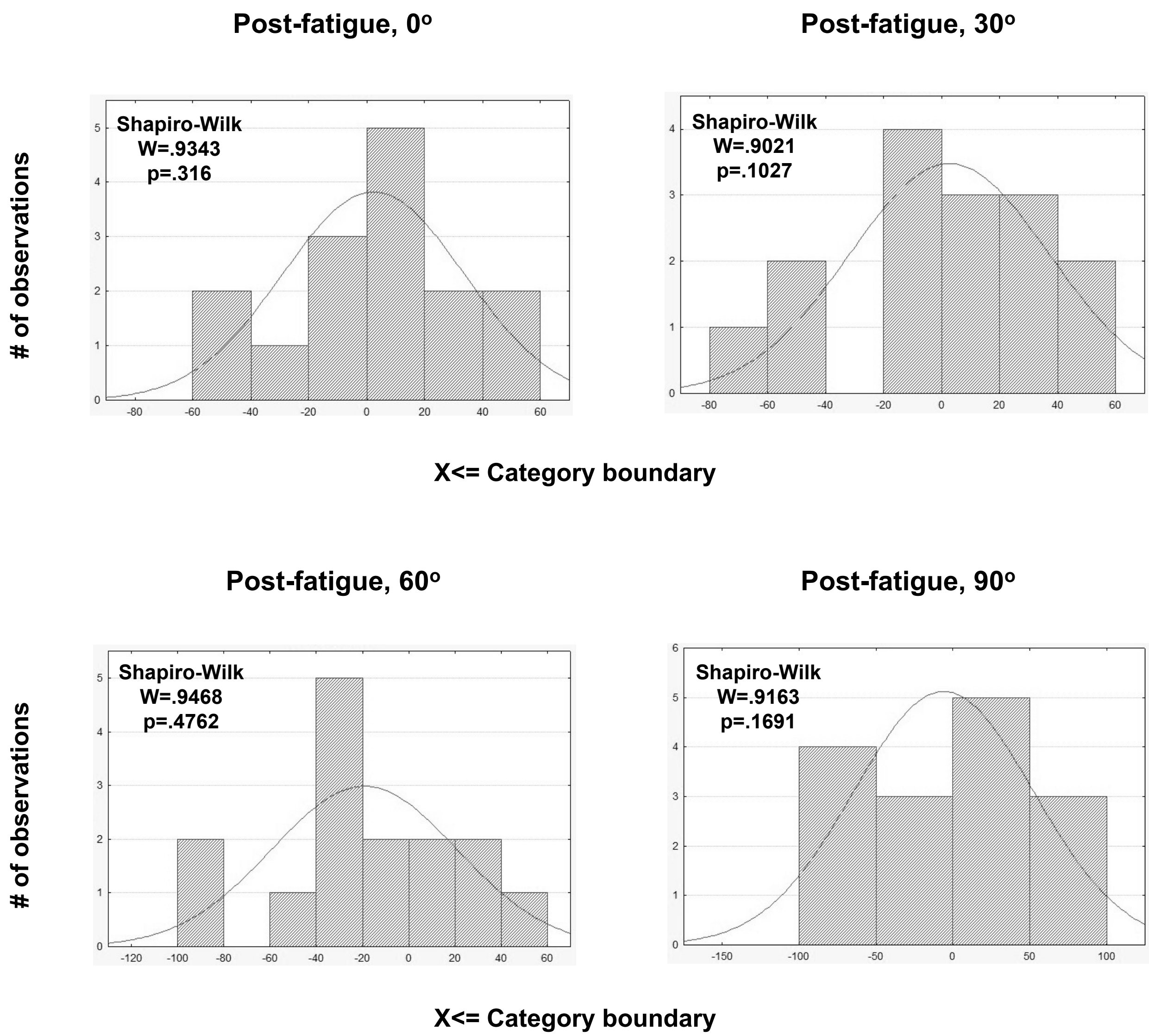


Figure B.4 Normality graphs for all participants for weight distribution in the sagittal plane for each knee position, after the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for antero-posterior (peaks) data

Pre-fatigue data (anterior peak)

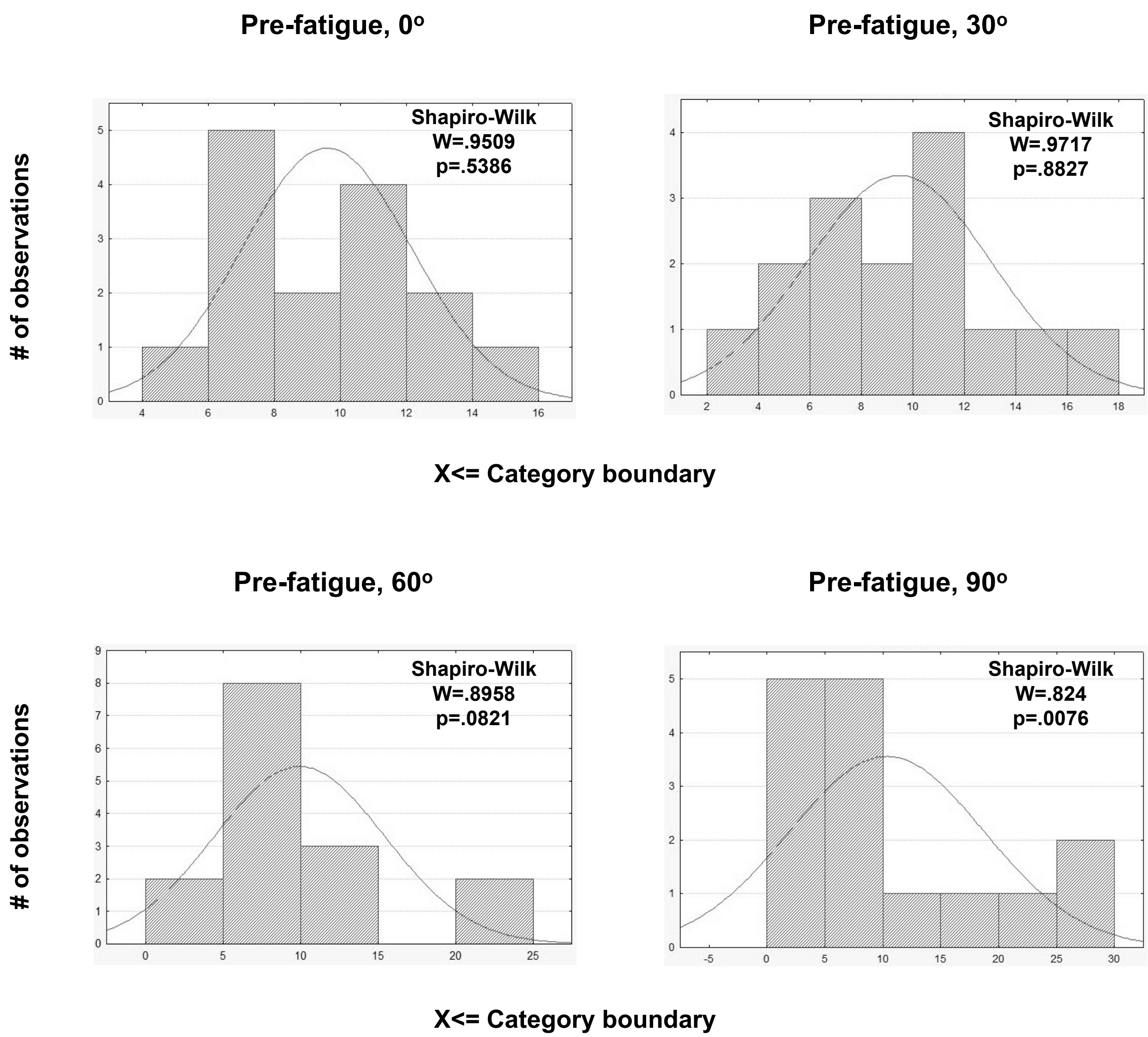


Figure B.5 Normality graphs for all participants for the amplitude of the anterior peak for each knee position, before the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for antero-posterior (peaks) data

Pre-fatigue data (posterior peak)

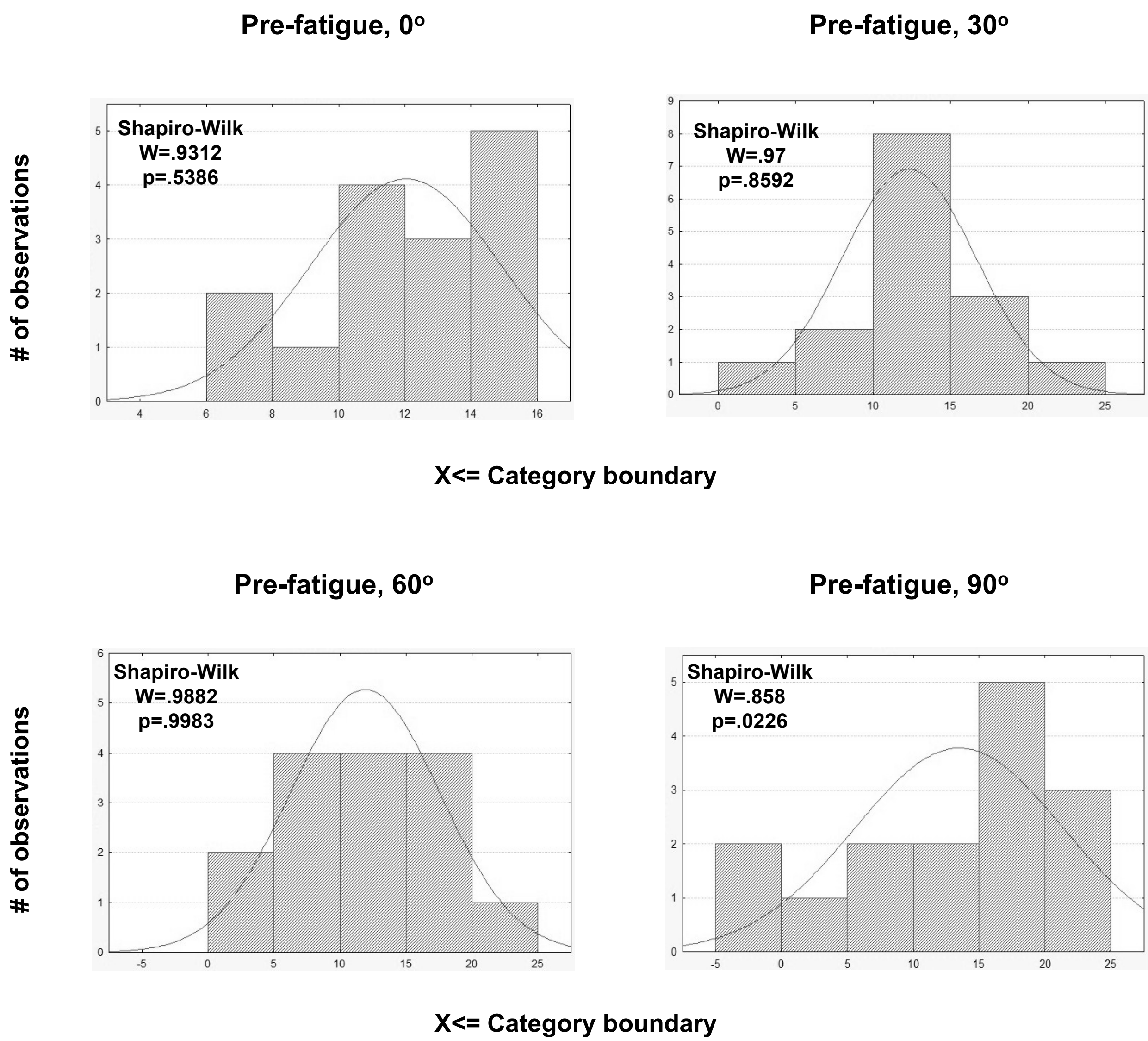


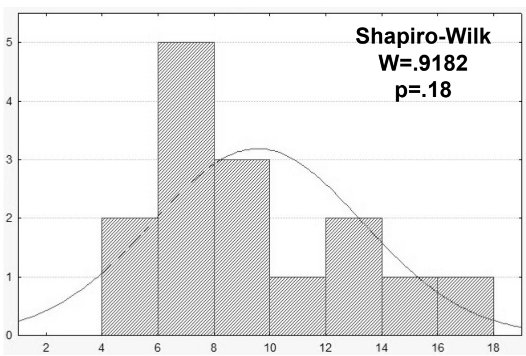
Figure B.6 Normality graphs for all participants for the amplitude of the posterior peak for each knee position, before the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for antero-posterior (peaks) data

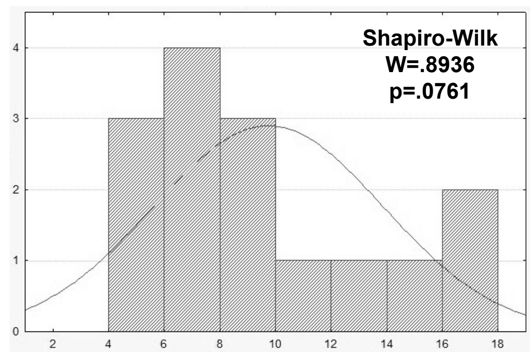
Post-fatigue data (anterior peak)

Post-fatigue, 0°

# of observations



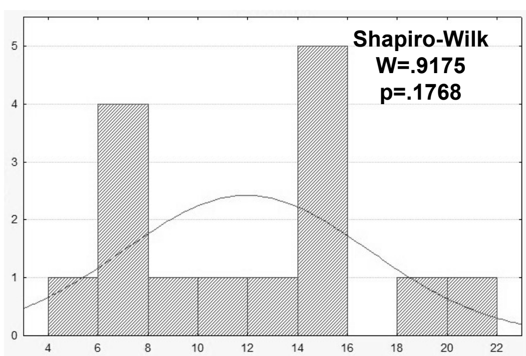
Post-fatigue, 30°



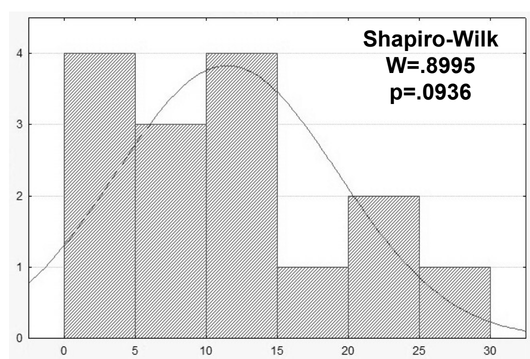
X<= Category boundary

Post-fatigue, 60°

# of observations



Post-fatigue, 90°



X<= Category boundary

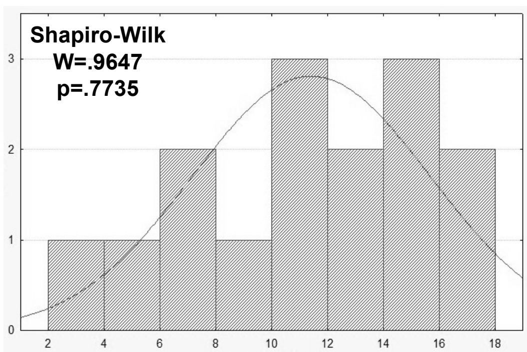
Figure B.7 Normality graphs for all participants for the amplitude of the anterior peak for each knee position, after the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for antero-posterior (peaks) data

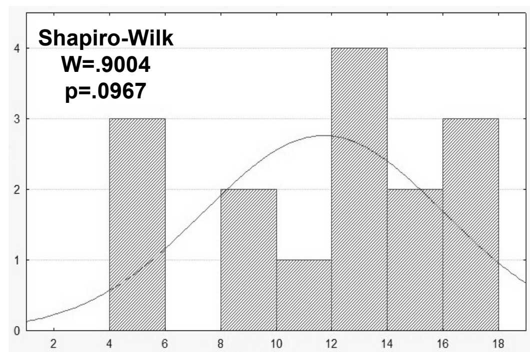
Post-fatigue data (posterior peak)

of observations

Post-fatigue, 0°



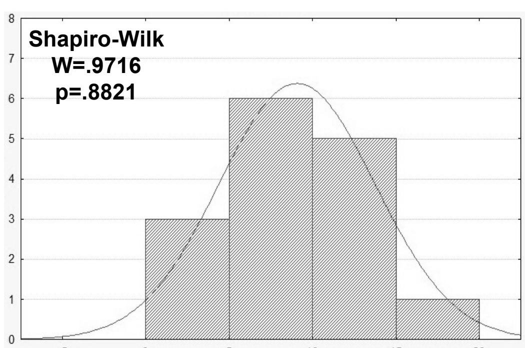
Post-fatigue, 30°



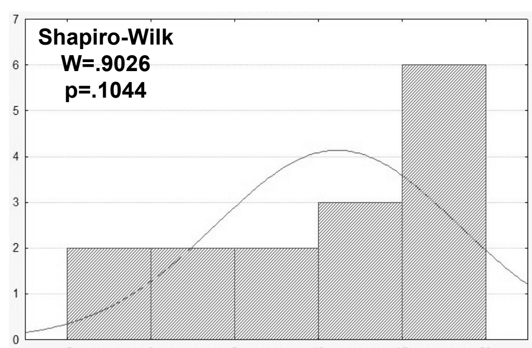
X<= Category boundary

# of observations

Post-fatigue, 60°



Post-fatigue, 90°



X<= Category boundary

Figure B.8 Normality graphs for all participants for the amplitude of the posterior peak for each knee position, after the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for COP position

Pre-fatigue data

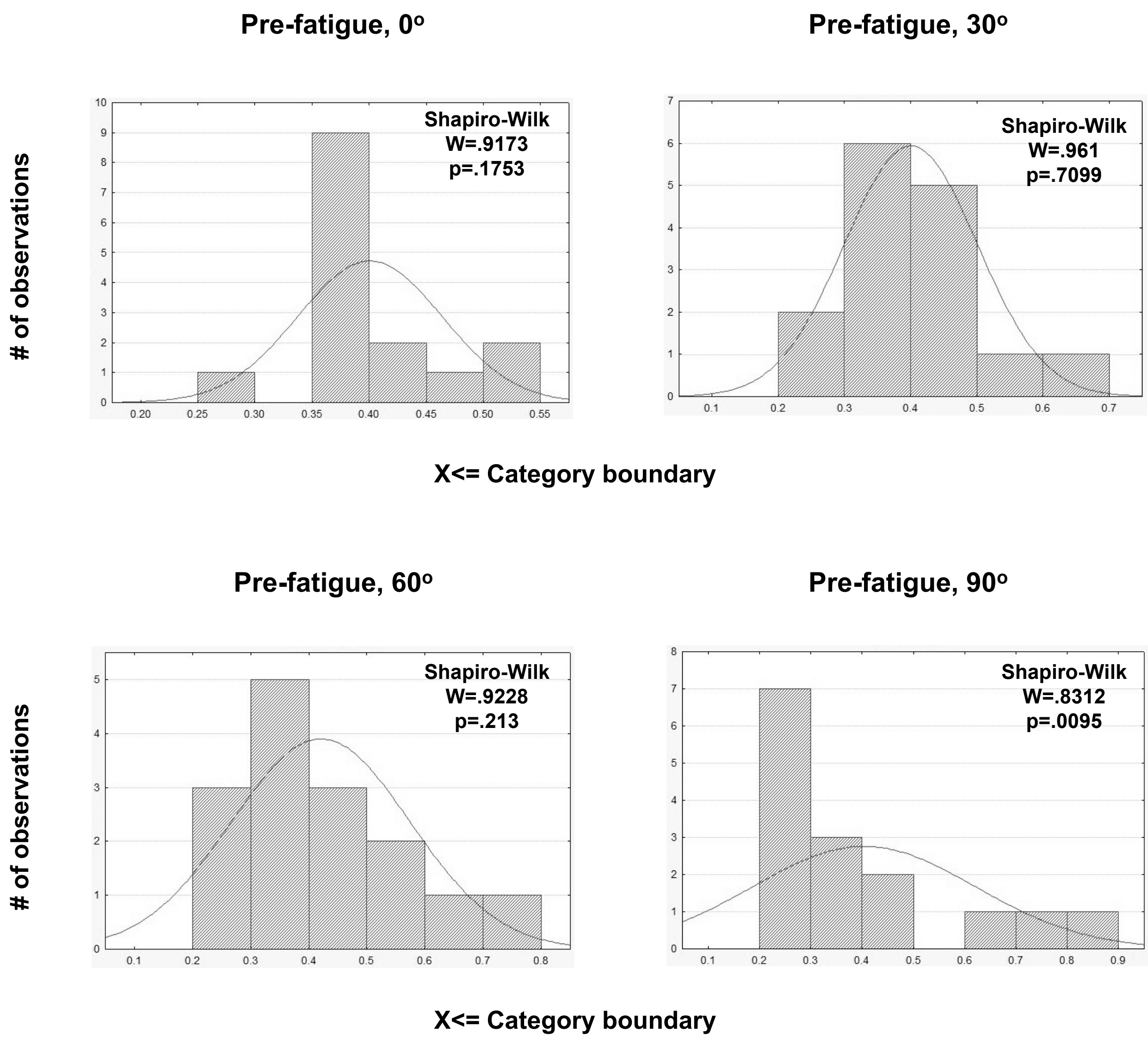


Figure B.9 Normality graphs for all participants for the COP position for each knee position, before the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Normality graphs for COP position

Post-fatigue data

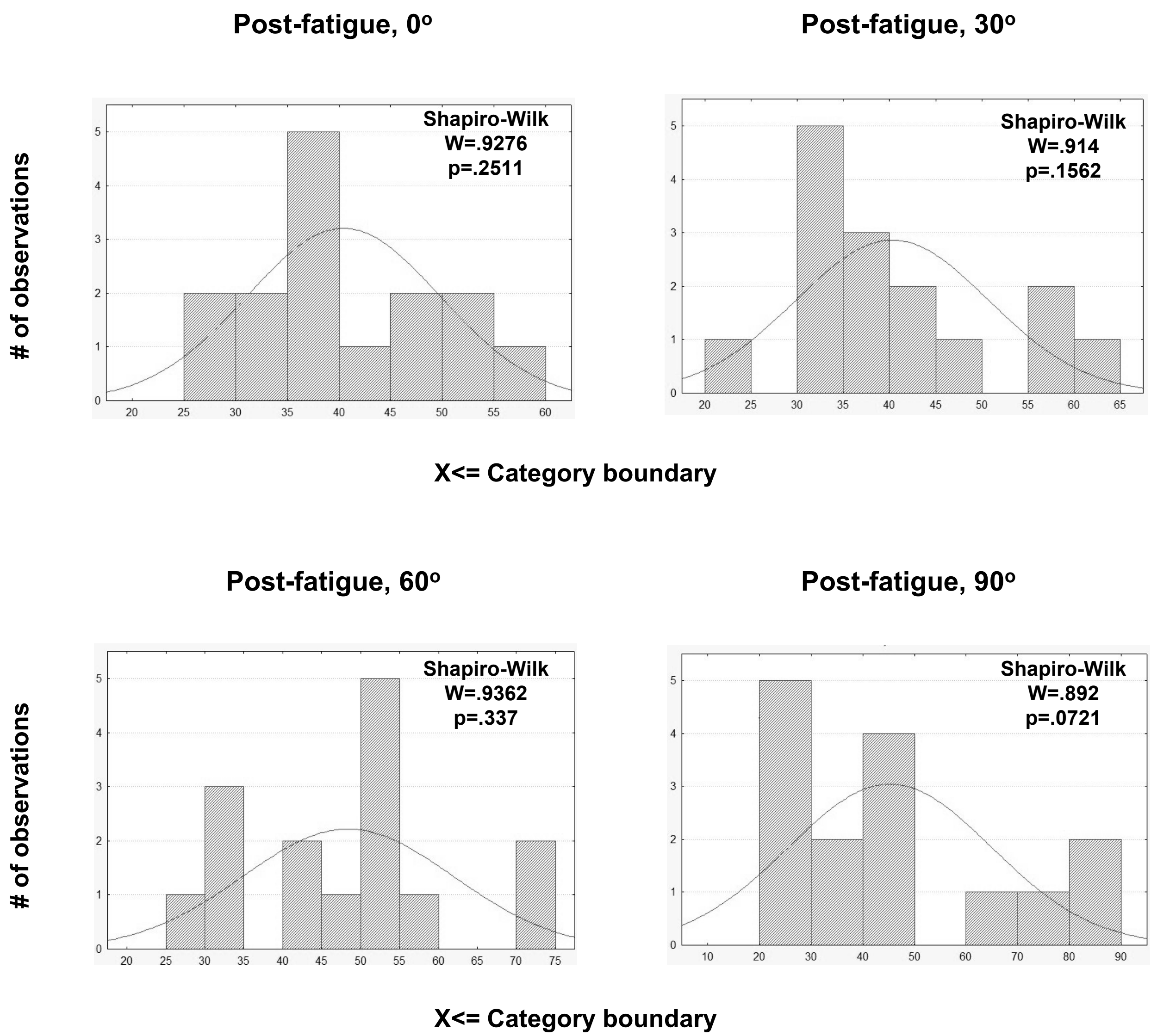


Figure B.10 Normality graphs for all participants for the COP position for each knee position, after the fatigue protocol. The W and the p values of the Shapiro-Wilk normality test are shown on each panel.

Appendix C

***Post-hoc* analysis for antero-posterior (peaks) weight distribution**

Table C.1.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		0.858651	0.989316	**0.000442**
30˚			0.963770	**0.003336**
60˚				**0.000910**

Table C.1.b. *Post-hoc* for main effect of fatigue

fatigue	*pre*	*post*
pre		**0.000911**

Table C.1.c. *Post-hoc* for peak*fatigue interaction

peak * fatigue	*pre anterior*	*post anterior*	*pre posterior*	*post posterior*
pre anterior		**0.010542**	0.633849	0.505820
post anterior			0.097324	0.141377
pre posterior				0.996184

P values are shown for *post hoc* analysis for antero-posterior (peak) weight distribution.

Significant p values (p<.05) are bolded

***Post-hoc* analysis for COP position**

Table C.2.a. *Post-hoc* for main effect of fatigue

fatigue	*pre*	*post*
pre		**0.039127**

Table C.2.b. *Post-hoc* for position*fatigue interaction

angle*fatigue	*pre 0˚*	*post0˚*	*pre30˚*	*post30˚*	*pre60˚*	*post60˚*	*pre90˚*	*post90˚*
pre 0˚		0.999994	1.000000	1.000000	0.831719	**0.000143**	0.999998	**0.011869**
post0˚			0.999992	1.000000	0.935340	**0.000156**	1.000000	**0.024612**
pre30˚				1.000000	0.825747	**0.000142**	0.999997	**0.011478**
post30˚					0.901456	**0.000149**	1.000000	**0.018544**
pre60˚						**0.001248**	1.000000	0.319654
post60˚							**0.000153**	0.350494
pre90˚								**0.350494**

P values are shown for *post hoc* analysis for center of pressure position.

Significant p values (p<.05) are bolded

Post-hoc analysis for rectus femoris muscle activity

Table C.3.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		**0.004130**	**0.000171**	**0.000171**
30˚			**0.003053**	**0.000171**
60˚				**0.000171**

Table C.3.b. *Post-hoc* for main effect of fatigue

fatigue	*pre*	*post*
pre		**0.002264**

Table C.3.c. *Post-hoc* for position*fatigue interaction

position*fatigue	*pre0˚*	*post0˚*	*pre30˚*	*post30˚*	*pre60˚*	*post60˚*	*pre90˚*	*post90˚*
pre0˚		1.000000	0.111755	**0.000524**	**0.000146**	**0.000135**	**0.000135**	**0.000135**
post0˚			0.092231	**0.000426**	**0.000143**	**0.000135**	**0.000135**	**0.000135**
pre30˚				0.480380	0.072459	**0.000135**	**0.000135**	**0.000135**
post30˚					0.970017	**0.000390**	**0.000135**	**0.000135**
pre60˚						**0.006087**	**0.000135**	**0.000135**
post60˚							**0.000135**	**0.000135**
pre90˚								**0.000135**

P values are shown for *post hoc* analysis for rectus femoris muscle activity.

Significant p values (p<.05) are bolded

Post-hoc analysis for vastus medialis muscle activity

Table C.4.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		**0.000174**	**0.000171**	**0.000171**
30˚			**0.000171**	**0.000171**
60˚				**0.000171**

Table C.4.b. *Post-hoc* for main effect of fatigue

fatigue	*pre*	*post*
pre		**0.000804**

Table C.4.c. *Post-hoc* for position*fatigue interaction

position*fatigue	*pre0˚*	*post0˚*	*pre30˚*	*post30˚*	*pre60˚*	*post60˚*	*pre90˚*	*post90˚*
pre0˚		0.999997	**0.000135**	**0.000135**	**0.000135**	**0.000135**	**0.000135**	**0.000135**
post0˚			**0.000135**	**0.000135**	**0.000135**	**0.000135**	**0.000135**	**0.000135**
pre30˚				0.094683	**0.000135**	**0.000135**	**0.000135**	**0.000135**
post30˚					**0.000135**	**0.000135**	**0.000135**	**0.000135**
pre60˚						**0.000776**	**0.000135**	**0.000135**
post60˚							**0.000135**	**0.000135**
pre90˚								**0.000135**

P values are shown for *post hoc* analysis for vastus medialis muscle activity

. Significant p values (p<.05) are bolded

***Post-hoc* analysis for tibialis anterior muscle activity**

Table C.5.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		0.958124	**0.003480**	**0.000171**
30˚			**0.013653**	**0.000171**
60˚				**0.000171**

P values are shown for *post hoc* analysis for tibialis anterior muscle activity.

Significant p values (p<.05) are bolded

***Post-hoc* analysis for biceps femoris muscle activity**

Table C.6.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		0.103513	0.140547	**0.020518**
30˚			0.998851	**0.000184**
60˚				**0.000193**

Table C.6.b. *Post-hoc* for main effect of fatigue

fatigue	*pre*	*post*
pre		**0.000462**

P values are shown for *post hoc* analysis for biceps femoris muscle activity.

Significant p values (p<.05) are bolded

***Post-hoc* analysis for semitendinosus muscle activity**

Table C.7.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		**0.013106**	**0.000490**	**0.000171**
30˚			0.603768	**0.000208**
60˚				**0.002275**

Table C.7.b. *Post-hoc* for position*fatigue interaction

position*fatigue	*pre0˚*	*post0˚*	*pre30˚*	*post30˚*	*pre60˚*	*post60˚*	*pre90˚*	*post90˚*
pre0˚		0.998222	**0.000611**	0.959205	**0.003980**	**0.010803**	**0.000138**	**0.000135**
post0˚			**0.003537**	0.999841	**0.023002**	0.056202	**0.000167**	**0.000135**
pre30˚				**0.012233**	0.997318	0.967533	0.844481	0.053048
post30˚					0.068984	0.150588	**0.000289**	**0.000135**
pre60˚						0.999963	0.437318	**0.009017**
post60˚							0.247837	**0.003303**
pre90˚								0.656866

P values are shown for *post hoc* analysis for semitendinosus muscle activity.

Significant p values (p<.05) are bolded

Post-hoc analysis for gastrocnemius lateralis muscle activity

Table C.8.a. *Post-hoc* for main effect of fatigue

fatigue	*pre*	*post*
pre		**0.015557**

P values are shown for *post hoc* analysis for gastrocnemius lateralis muscle activity.

Significant p values (p<.05) are bolded

Post-hoc analysis for gastrocnemius medialis muscle activity

Table C.9.a. *Post-hoc* for main effect of position

position	*0˚*	*30˚*	*60˚*	*90˚*
0˚		**0.007550**	**0.002843**	0.057955
30˚			0.984698	0.851394
60˚				0.655391

P values are shown for *post hoc* analysis for gastrocnemius medialis muscle activity.

Significant p values ($p<.05$) are bolded

Appendix D

Table D.1. Non-dominant - dominant weight distribution differences data across participants

	pre-fatigue				post-fatigue			
	0˚	30˚	60˚	90˚	0˚	30°	60˚	90˚
CTL01	(+)	(+)	(+)	(n)	(+)	(+)	(n)	(n)
CTL02	(n)	(n)	(+)	(+)	(n)	(n)	(n)	(+)
CTL03	(+)	(+)	(n)	(+)	(+)	(n)	(-)	(+)
CTL04	(+)	(+)	(+)	(+)	(+)	(+)	(n)	(+)
CTL05	(n)	(-)	(n)	(+)	(n)	(n)	(n)	(+)
CTL06	(n)	(-)	(+)	(+)	(-)	(+)	(+)	(+)
CTL07	(-)	(-)	(-)	(-)	(-)	(-)	(-)	(-)
CTL08	(-)	(-)	(-)	(n)	(-)	(-)	(-)	(-)
CTL09	(n)	(n)	(+)	(n)	(n)	(-)	(-)	(-)
CTL10	(+)	(+)	(+)	(+)	(+)	(+)	(+)	(n)
CTL11	(n)	(+)	(+)	(+)	(-)	(+)	(n)	(n)
CTL12	(n)	(+)	(+)	(n)	(n)	(+)	(+)	(n)
CTL13	(-)	(-)	(n)	(-)	(n)	(-)	(n)	(n)
CTL14	(+)	(+)	(n)	(n)	(n)	(n)	(-)	(+)
CTL15	(+)	(n)	(n)	(n)	(+)	(+)	(n)	(+)

(+): positive, larger than 3, more weight on the non-dominant

(-): negative, smaller than -3, more weight on the dominant

(n): neutral, between -3 and 3

Table D.2. Antero-posterior weight distribution differences data across participants

	pre-fatigue				post-fatigue			
	0˚	30˚	60˚	90˚	0˚	30˚	60˚	90˚
CTL01	(+)	(+)	(+)	(+)	(n)	(+)	(n)	(+)
CTL02	(+)	(+)	(+)	(+)	(+)	(+)	(+)	(+)
CTL03	(n)	(-)	(-)	(-)	(-)	(-)	(-)	(-)
CTL04	(+)	(n)	(-)	(-)	(+)	(n)	(-)	(-)
CTL05	(-)	(n)	(+)	(+)	(-)	(-)	(-)	(n)
CTL06	(-)	(n)	(n)	(-)	(-)	(+)	(+)	(-)
CTL07	(+)	(+)	(+)	(+)	(+)	(+)	(-)	(n)
CTL08	(-)	(+)	(+)	(+)	(-)	(+)	(+)	(+)
CTL09	(n)	(-)	(-)	(-)	(+)	(n)	(-)	(-)
CTL10	(n)	(n)	(-)	(-)	(+)	(+)	(-)	(-)
CTL11	(n)	(+)	(n)	(n)	(n)	(+)	(n)	(n)
CTL12	(n)	(+)	(+)	(+)	(+)	(+)	(+)	(+)
CTL13	(n)	(-)	(-)	(n)	(n)	(n)	(-)	(-)
CTL14	(n)	(-)	(n)	(+)	(-)	(-)	(-)	(+)
CTL15	(n)	(+)	(+)	(+)	(n)	(n)	(-)	(+)

(+): positive, larger than 15, more weight posteriorly
(-): negative, smaller than -15, more weight anteriorly
(n): neutral, between -15 and 15

Table D.3. COP position data across participants

	pre-fatigue				post-fatigue			
	0˚	30˚	60˚	90˚	0˚	30˚	60˚	90˚
CTL01	↓	↓	↓	↓	~	↓	~	↓
CTL02	~	↓	↓	↓	↓	↓	↓	↓
CTL03	~	↑	↑	↑	↑	↑	↑	↑
CTL04	↓	~	↑	↑	↓	↑	↑	↑
CTL05	↑	~	↓	↓	↑	↑	↑	~
CTL07	↑	~	↓	↑	↑	↓	↓	↑
CTL08	~	↓	~	↓	~	↓	↑	~
CTL09	↑	↓	↓	↓	↑	↓	↓	↓
CTL11	~	↑	↑	↑	↓	~	↑	↑
CTL12	~	↑	↑	↑	↓	~	↑	↑
CTL14	~	↓	~	~	~	↓	~	~
CTL16	~	↓	~	↓	~	↓	↓	↓
CTL17	~	↑	↑	~	~	~	↑	↑
CTL18	~	↑	~	↓	↑	↑	↑	↓
CTL19	~	↓	~	↓	~	~	↑	↓

↑: larger than 0.45, COP position closer to the toes
↓: smaller than 0.35, COP position closer to the heels
~: between 0.35 and 0.45

Table D.4. Antero-posterior peaks ratio across participants
Posterior peak/anterior peak

	pre-fatigue				post-fatigue			
	0˚	30˚	60˚	90˚	0˚	30˚	60˚	90˚
CTL01	↑	↑	↑	↑	↑	↑	↑	↑
CTL02	↑	↑	↑	↑	↑	↑	↑	↑
CTL03	~	↓	↓	↓	↓	↓	↓	↓
CTL04	↑	~	↓	↓	↑	↓	↓	↓
CTL05	↓	~	↑	↑	↓	↓	↓	↑
CTL06	↓	~	↑	↓	↓	↑	↑	↓
CTL07	↑	↑	↑	↑	↑	↑	↓	~
CTL08	↓	↑	↑	↑	↓	↑	↑	↑
CTL09	↑	~	↓	↓	↑	~	↓	↓
CTL10	↑	~	↓	↓	↑	↑	↓	↓
CTL11	↑	↑	↑	↑	↑	↑	~	↑
CTL12	↑	↑	↑	↑	↑	↑	↑	↑
CTL13	↑	~	↓	↑	↑	~	↓	↓
CTL14	↑	~	~	↑	↓	↓	↓	↑
CTL15	↑	↑	↑	↑	↑	↑	~	↑

↑: larger than 1.2, larger amplitude for the posterior peak
↓: smaller than 0.8, larger amplitude for the anterior peak
~: between 0.8 and 1.2

Table D.5. EMG RMS activity ratios (right/left)
Pre-fatigue

		RF	VM	TA	BF	ST	GL	GM
CTL01	0˚	~	↑	↓	↑	~	↑	↓
	30˚	↓	↓	↓	~	↓	~	↓
	60˚	↑	↓	~	~	↓	~	~
	90˚	~	~	~	~	~	~	~
CTL02	0˚	↓	↑	↑	↓	↑	~	↑
	30˚	↓	~	↓	↑	↓	↓	↓
	60˚	↓	↓	~	~	↓	~	~
	90˚	~	~	~	~	~	~	~
CTL03	0˚	↑	~	↑	↓	~	↑	~
	30˚	↑	↓	↑	~	~	↑	↑
	60˚	~	~	↑	~	~	↑	~
	90˚	~	~	~	~	~	~	~
CTL04	0˚	↓	↑	↓	↓	↓	↑	↓
	30˚	↓	↓	~	↓	↓	↑	↓
	60˚	↓	~	↓	~	~	↑	↓
	90˚	~	~	~	~	~	~	~
CTL05	0˚	↓	↓	↑	↑	↓	↓	↓
	30˚	↑	↑	~	↓	↓	↓	↓
	60˚	↑	~	~	↑	↑	~	↓
	90˚	~	~	~	~	~	~	~
CTL06	0˚	~	↑	~	↓	↓	~	↓
	30˚	↑	↑	~	↓	↓	↓	↓
	60˚	↑	↑	↓	↑	~	↑	↑
	90˚	~	~	~	~	~	~	~
CTL07	0˚	↑	↑	↓	↓	↓	↓	↓
	30˚	↑	↑	~	↓	↓	↓	↓
	60˚	↑	↑	↑	~	↑	~	~
	90˚	~	~	~	~	~	~	~
CTL08	0˚	~	~	~	↓	↑	↓	~
	30˚	↑	↑	~	↑	↑	~	↑
	60˚	~	↑	↑	~	↑	↑	↑
	90˚	~	~	~	~	~	~	~

↑: larger than 1.2, more activity on the right side;
↓: smaller than 0.8, more activity on the left side;
~: between 0.8 and 1.2

Table D.5. EMG RMS activity ratios (right/left) – cont'd
Pre-fatigue

		RF	VM	TA	BF	ST	GL	GM
CTL09	0˚	↑	↑	↓	↓	↑	↓	↓
	30˚	↑	~	↓	↓	↑	↓	↓
	60˚	~	~	~	~	~	↓	↓
	90˚	~	~	~	~	~	~	~
CTL10	0˚	↑	↑	~	↑	↑	~	~
	30˚	↓	↑	↑	~	↑	~	↓
	60˚	~	↑	~	~	↑	↓	↓
	90˚	~	~	~	~	~	~	~
CTL11	0˚	↑	↑	↓	~	~	↓	~
	30˚	~	↑	↑	↑	↓	↓	↑
	60˚	~	~	↑	↑	~	↓	~
	90˚	~	~	~	~	~	~	~
CTL12	0˚	↑	↓	~	↑	↑	~	↓
	30˚	↑	~	↑	↓	~	↑	~
	60˚	↑	↓	↑	↓	↓	~	~
	90˚	~	~	~	~	~	~	~
CTL13	0˚	↓	↓	↑	↑	↓	~	↑
	30˚	~	~	↓	~	↓	~	~
	60˚	↑	~	~	~	↓	↑	↑
	90˚	~	~	~	~	~	~	~
CTL14	0˚	↑	↑	↑	↑	~	~	~
	30˚	~	~	↑	↓	↓	~	~
	60˚	~	~	~	~	↓	~	~
	90˚	~	~	~	~	~	~	~
CTL15	0˚	~	↑	↓	↑	↑	↑	↑
	30˚	↓	~	↓	↑	↑	~	↑
	60˚	↓	~	~	~	~	~	↑
	90˚	~	~	~	~	~	~	~

↑: larger than 1.2, more activity on the right side;
↓: smaller than 0.8, more activity on the left side;
~: between 0.8 and 1.2

Table D.6. EMG RMS activity ratios (right/left)
Post-fatigue

		RF	VM	TA	BF	ST	GL	GM
CTL01	0˚	↑	↓	~	↑	↑	↑	~
	30˚	~	↓	↑	↓	↓	↓	↓
	60˚	↑	~	↑	~	↓	↑	↓
	90˚	↑	~	~	~	~	~	↓
CTL02	0˚	↓	↓	↑	↓	↑	~	↑
	30˚	↓	↓	↓	↓	~	↑	~
	60˚	↑	~	~	~	↓	~	~
	90˚	↑	↑	↑	~	~	↑	↑
CTL03	0˚	↑	~	↑	↓	~	↑	↑
	30˚	~	↓	↑	↓	↓	↑	↑
	60˚	↑	~	↑	~	~	↑	↑
	90˚	~	↑	↑	~	~	~	~
CTL04	0˚	↓	~	↓	~	↓	↑	↓
	30˚	↓	↓	~	↑	↓	↑	↓
	60˚	↓	↓	↑	↑	~	↑	↓
	90˚	↓	~	↑	↑	↑	↑	↓
CTL05	0˚	↑	↑	~	↑	↑	↓	↑
	30˚	↑	↑	~	↓	↑	↓	↓
	60˚	↑	↑	~	~	↑	↓	↓
	90˚	↑	↑	↑	↑	~	~	~
CTL06	0˚	↑	~	↑	↓	↓	~	↓
	30˚	~	↑	↓	↓	↓	↓	↓
	60˚	~	↑	↓	↑	↑	↑	↑
	90˚	~	~	↑	~	~	↓	~
CTL07	0˚	↓	↓	↓	↓	↓	↓	↓
	30˚	↑	↑	~	~	↑	↓	↓
	60˚	↑	↑	↑	~	↑	↓	↓
	90˚	↑	↑	~	~	~	↓	↓
CTL08	0˚	~	~	~	↓	↑	~	↓
	30˚	~	↑	~	↑	↑	↑	↑
	60˚	~	↑	↓	~	~	↓	↓
	90˚	~	~	~	~	~	~	~

↑: larger than 1.2, more activity on the right side;
↓: smaller than 0.8, more activity on the left side;
~: between 0.8 and 1.2

Table D.6 EMG RMS activity ratios (right/left) – cont'd
Post-fatigue

		RF	VM	TA	BF	ST	GL	GM
CTL09	0˚	↑	~	↓	↓	↓	↓	↓
	30˚	↑	↑	↓	↓	↓	↓	↓
	60˚	~	~	↓	~	↑	↓	~
	90˚	↑	↓	↓	~	↑	↓	~
CTL10	0˚	↑	~	↑	↑	↑	↑	↑
	30˚	↑	~	↓	~	↑	~	~
	60˚	↑	↑	~	↑	↑	~	~
	90˚	↑	↑	↑	~	↑	~	~
CTL11	0˚	↑	↑	↓	↑	↑	↓	~
	30˚	↑	↑	↑	↑	~	↓	↓
	60˚	↑	↑	↑	↑	↑	↓	↓
	90˚	↑	↑	~	↑	~	↓	↓
CTL12	0˚	↑	↓	↑	↑	↓	↑	↓
	30˚	↓	~	↑	↓	↓	↑	↑
	60˚	~	~	↑	~	~	~	~
	90˚	~	~	~	~	~	~	~
CTL13	0˚	↓	↓	~	↓	↓	~	↓
	30˚	↑	↑	~	~	↓	~	~
	60˚	↑	~	~	~	~	↑	~
	90˚	↑	↑	~	↑	↑	↑	↑
CTL14	0˚	↑	~	~	↑	↓	~	↓
	30˚	↑	↑	↑	↑	↓	~	~
	60˚	↑	↑	~	~	~	↑	↑
	90˚	↓	~	~	~	~	~	~
CTL15	0˚	~	↑	↓	↑	↑	↑	↑
	30˚	↓	~	↓	~	~	↑	↑
	60˚	↓	↓	~	~	~	↑	~
	90˚	↑	~	~	~	~	~	~

↑: larger than 1.2, more activity on the right side;
↓: smaller than 0.8, more activity on the left side;
~: between 0.8 and 1.2

Appendix E

Matlab script for the mathematical transformation of the antero-posterior weight distribution vectors

```
% Convert a weight distribution vectors into a 20-element vectors
% normalized to body weight

function AP_weightD(tag)

%subject to be analyzed
subject=tag(1:5);
session=tag(7:8);

pathmatlab=pwd;

cd('C:\Users\Nancy\Posture Lab\ACL - Fatigue project\Data\MatScan data (new
way)\')
cd('Controls')
cd(subject)
cd(session)

% load data (vector and sensor where front of foot not including toes begins)
% all trials (3 pre and 3 post)
eval(['load ' tag '_wdist_AP.txt'])
eval(['load ' tag '_feetline'])

eval(['inputmat=' tag '_wdist_AP;'])
eval(['vbegin=' tag '_feetline;'])


% convert each trial individually
for i=1:6
    biggermat=[];
    finalmat=[];
    normmat=[];

   % find front and back of feet (including toes)
    mat=inputmat(:,5*(i-1)+1:5*i-1);
    vnotzerobegin=(find(sum(mat')~=0))';
    vnotzeroend=flipud((find(sum(mat')~=0))');
    toes=mean(mat(vnotzerobegin(1):vbegin(i)-1,:));
    newmat=[toes;mat(vbegin(i):vnotzeroend(1),:)];
    len=vnotzeroend(1)-vbegin(i)+2;


    % convert into long vector (20*size)
    for j=1:len
        biggermat(20*(j-1)+1:20*j,1)=newmat(j,1);
        biggermat(20*(j-1)+1:20*j,2)=newmat(j,2);
        biggermat(20*(j-1)+1:20*j,3)=newmat(j,3);
        biggermat(20*(j-1)+1:20*j,4)=newmat(j,4);
    end

    % convert into 20-element vector
    for j=1:20
```

```
        finalmat(j,:)=mean(biggermat(len*(j-1)+1:len*j,:));
    end

    % normalize to body weight
    normmat(:,1)=100*finalmat(:,1)/sum(finalmat(:,1));
    normmat(:,2)=100*finalmat(:,2)/sum(finalmat(:,2));
    normmat(:,3)=100*finalmat(:,3)/sum(finalmat(:,3));
    normmat(:,4)=100*finalmat(:,4)/sum(finalmat(:,4));

    threeDmat(:,:,i)=normmat;

end

% split matrix into 6 trials
eval([tag '_pre_t1_apwd=threeDmat(:,:,1);']);
eval([tag '_pre_t2_apwd=threeDmat(:,:,2);']);
eval([tag '_pre_t3_apwd=threeDmat(:,:,3);']);
eval([tag '_post_t1_apwd=threeDmat(:,:,4);']);
eval([tag '_post_t2_apwd=threeDmat(:,:,5);']);
eval([tag '_post_t3_apwd=threeDmat(:,:,6);']);


% average 3 pre trials and 3 post trials
eval([tag '_pre_apwd=squeeze(mean(permute(threeDmat(:,:,1:3),[3,1,2])));']);
eval([tag '_post_apwd=squeeze(mean(permute(threeDmat(:,:,4:6),[3,1,2])));']);

% save trials
eval(['save ' tag '_APWD_pre_t1.txt ' tag '_pre_t1_apwd -ASCII'])
eval(['save ' tag '_APWD_pre_t2.txt ' tag '_pre_t2_apwd -ASCII'])
eval(['save ' tag '_APWD_pre_t3.txt ' tag '_pre_t3_apwd -ASCII'])
eval(['save ' tag '_APWD_post_t1.txt ' tag '_post_t1_apwd -ASCII'])
eval(['save ' tag '_APWD_post_t2.txt ' tag '_post_t2_apwd -ASCII'])
eval(['save ' tag '_APWD_post_t3.txt ' tag '_post_t3_apwd -ASCII'])

% save means (pre mean and post mean)
eval(['save ' tag '_APWD_pre.txt ' tag '_pre_apwd -ASCII'])
eval(['save ' tag '_APWD_post.txt ' tag '_post_apwd -ASCII'])

cd(pathmatlab)

clear
```