The Effect of Cutaneous Sensation, Proprioception and Strength in the Non-Amputated Leg on Balance in Traumatic Transtibial Amputees

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A Thesis

in

The Department

of

Exercise Science

Presented in Partial Fulfillment of the Requirements

for the Degree of Master of Science (Exercise Science) at

Concordia University

Montreal, Quebec, Canada

August 2017

CONCORDIA UNIVERSITY

School of Graduate Studies

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Entitled: **The Effect of Cutaneous Sensation, Proprioception and Strength in the Non-Amputated Leg on Balance in Traumatic Transtibial Amputees**

and submitted in partial fulfillment of the requirements for the degree of

Master of Science

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ABSTRACT

The Effect of Cutaneous Sensation, Proprioception and Strength in the Non-Amputated Leg on Balance in Traumatic Transtibial Amputees

Elyse Courville

As of 2005, there were 1.6 million people living with limb loss in the US. Studies have shown an effect of amputation on balance. Balance has mainly been studied during quiet standing in double-legged stance and results are conflicting regarding the amount of sway in amputees compared to controls. Few studies have assessed single-legged stance, but they all agree that sway is not different when standing on the non-amputated leg only or on a control leg, although standing time is decreased in amputees. Other variables, including cutaneous sensation, proprioception and strength have also been found to be decreased in both the amputated and non-amputated legs. Not much is known about single-legged stance, perturbed standing or the relationship between balance and cutaneous sensation, proprioception and strength in amputees. The goal of this study was to assess balance in traumatic transtibial amputees during perturbations when standing on the non-amputated leg. Whole body kinematics, 3D ground reaction forces and activity in eight trunk and lower limb muscles were recorded during the balance testing. Three other measurements were taken on the same leg as the balance test: cutaneous sensation, proprioception and strength.

There is no difference in sway between amputees and able-bodied controls although there is a trend suggesting an increase in sway in controls after the perturbation. Amputees exhibited less variability than controls in sway variables. Decreased variability in the balance task may be due to decreased balance confidence in amputees. No differences were found between groups in cutaneous sensation, proprioception and strength although a trend was revealed for controls to be stronger than amputees. Although some aspects of balance were significantly correlated to cutaneous sensation and proprioception, strength was the measure that was most often significantly correlated with balance in amputated individuals. Electromyographic activity increased after the perturbation in five muscles, although no differences were found between groups. The amputees and controls also did not differ in terms of muscle latencies, but a different activation pattern was found in the peroneus longus.

ACKNOWLEDGEMENTS

I must thank first and foremost my supervisor Dr. Nancy St-Onge for providing guidance and support throughout this project. I am lucky for having had such an inspiring supervisor. I will always cherish our thought-provoking and rewarding problem-solving sessions. Her enthusiasm is contagious, it instilled the research bug in me and reinforced my critical thinking. Thank you for reviewing my many drafts so thoroughly and affording me so much of your time.

Thank you to the department of Exercise Science for the funding I have received over the years to make this project possible, including the PERFORM seed funding from Dr. Nancy St-Onge. I am grateful to have received the Graduate Merit Scholarship – Faculty of Arts and Science as well as the William R. Sellers Award. The opportunity to teach and assist labs and lectures in the department not only afforded more funding but also provided me with valuable experience and a love for teaching.

Thank you to the PERFORM Center for providing the essential equipment, labs and support staff that made this research possible and to the Institut de Réadaptation Gingras-Lindsay-de-Montréal that enabled the recruitment process. Thank you to all of the participants who accepted to take part in this study.

A special thank you goes out to the three undergraduate students who helped me with recruitment, data collection, data analysis, and putting a smile on my face. Thank you to Kristina, Matylda and Michael.

Thank you to my committee members, Dr. Alain Leroux and Dr. Shawn Robbins for providing me with valuable feedback that served to help improve this project. A special thank you to Dr. Geoffrey Dover for providing feedback and direction for the strength data. I also cannot forget to thank Dr. Robert Kilgour whose enthusiasm for research and learning is the reason I started this Master's.

Thank you to the faculty, support staff, graduate and undergraduate students of the department of Exercise Science, for providing such a wonderful work environment.

Merci à la meilleure famille au monde, à mes parents et à mon frère. Thank you to the best friends a girl could ask for. I am so lucky and am grateful for my amazing support system.

CONTRIBUTION OF AUTHORS

For the manuscript titled: Traumatic Transtibial Amputees Display Tight Control of Center of Pressure in Response to Perturbations

Elyse Courville: participated in protocol development, conducted recruitment, data collection, data analysis, Matlab scripts, statistical analysis, interpretation of data and preparation of manuscript

Nancy St-Onge: developed the protocol, supervised data collection, analysis, Matlab scripts, statistical analysis, interpretation of data and preparation of manuscript.

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LIST OF ABBREVIATIONS

AAS: Adjusted Activity Score

AP: Anterior-Posterior

BF: Biceps Femoris

COM: Center of Mass

COP: Center of Pressure

DF/PF: Dorsiflexors/Plantarflexors

EMG: Electromyography

ES: Erector Spinae

F/E: Flexors/Extensors

GM: Gastrocnemius Medialis

HAP: Human Activity Profile

JR: Joint Repositioning

MAS: Maximum Activity Score

ML: Medio-Lateral

PL: Peroneus Longus

PT/BW: Peak Torque to Body Weight

RA: Rectus Abdominis

RF: Rectus Femoris

RMS: Root Mean Square

TA: Tibialis Anterior

TDPM: Threshold Detection of Passive Motion

TPS: Touch Pressure Sensation

VM: Vastus Medialis

CHAPTER 1 − THEORETICAL CONTEXT

Balance is needed for activities of daily living, and it is composed of many elements. If one of these elements is not functioning properly then balance may be affected. In lower-limb amputees, muscles and tissues are missing, potentially affecting some of the elements. The elements of interest in this study are cutaneous sensation, proprioception and strength. The goal of this study is to assess if and how balance is affected in amputees and what influence the different elements have on balance. Previous studies have looked at center of pressure (COP) displacement to assess the degree to which balance is affected. Correlations between cutaneous sensation, proprioception, strength and balance have rarely been assessed in amputees. Muscle activation has also been studied to explain its contribution to balance control in amputees.

1.1 Epidemiology

According to Ownings and Kozak (1998), it is estimated that 185,000 people have a new upper or lower limb amputation every year in the United States. As of 2005, there were 1.6 million people living with limb loss in the US and this number is expected to double by 2050, following recent trends (Ziegler-Graham et al., 2008). Ziegler-Graham et al. (2008) report that 65% of amputations are lower limb amputations. In lower limb amputations, transtibial amputations are the most common (Kayssi et al., 2016a; Pezzin et al., 2000). Vascular problems (54%), including diabetes, are the leading cause of amputation followed by trauma (45%) and then cancer (<2%) (Ebskov, 1992; Gregory-Dean, 1991; Kayssi et al., 2016a; Kayssi et al., 2016b; Moxey et al., 2010; Ziegler-Graham et al., 2008). Sixty-four percent of vascular amputations occur in people over the age of 65 (Ziegler-Graham et al., 2008). Traumatic amputations generally affect people under 60 years old with traffic accidents as the leading cause (Dillingham et al., 1998; Gregory-Dean, 1991; Pezzin et al., 2000). Indeed, two-thirds of traumatic amputees are under the age of 45 at the time of amputation (Ziegler-Graham et al., 2008). Approximately 86-90% of major traumatic amputations affect males (Dillingham et al., 1998; Pezzin et al., 2000). Gregory-Dean (1991) states that amputations in general are 2.5 times more likely to occur in males compared to females, but that this number evens out with age.

1.2 Cutaneous Sensation

The sense of touch is composed of different elements. For example, light touch and deep pressure can be distinguished from one another. Vibrations as well as superficial pain can be felt through the skin. There are many different types of receptors in the skin such as mechanoreceptors, thermoreceptors and nociceptors (Purves et al., 2004). Each receptor will perceive a different sensation, for example, mechanoreceptors will perceive light touch, vibrations, and deep pressure (Purves et al., 2004). Seeing as amputees are missing certain tissues, cutaneous sensation could be affected.

Most cutaneous sensation tests have shown a change in sensation in the amputated limb. The non-amputated limb is less frequently tested and often serves as a control for the amputated leg. When the non-amputated leg is tested, it has been found that there is a decrease in cutaneous sensation compared to able-bodied control legs, regardless of the testing protocol.

To our knowledge, only one group of researchers looked at Touch Pressure Sensation (TPS) which determines the smallest size monofilament that can be felt and returns a value in log 10 mg (Kavounoudias et al., 2005). Monofilaments are first used in ascending order to establish the smallest sized monofilament that can be felt. The test is then repeated in descending order to determine the first filament that is not felt anymore. Kavounoudias et al. (2005) had separate groups for vascular and traumatic amputees all of which were transtibial amputees. Kavounoudias et al. (2005) tested TPS on the anterior surface of the tibia and the plantar aspect of the foot and found that in the traumatic transtibial group TPS values were larger on the tibial site of the amputated leg compared to non-amputated leg, meaning there was a decrease in sensation in the amputated leg. In the vascular group though, there was no difference in TPS values between the amputated and non-amputated leg at the tibial site. When comparing the non-amputated leg to that of the able-bodied control group, there was no difference in TPS values at the tibial site for either group of amputees. The traumatic group had higher TPS values on the plantar aspect of the non-amputated leg than their respective able-bodied, age-matched control groups. The difference in TPS values at the plantar site between vascular amputees and controls was not significant. It is speculated that the reason there is a decrease in the cutaneous sensation in the plantar aspect of the non-amputated leg in traumatic amputees is that there is a reorganization that decreases the sensation to match that of the amputated limb (Kavounoudias et al., 2005). When comparing the tibial and plantar sites, in both amputee groups, it was found that the tibial site of the non-amputated leg had a lower TPS value than the foot of the same leg.

The accuracy of a person's sense of touch can be determined through many tests. Some tests do not produce a precise value like the TPS test, but rather the participant must respond whether they feel the stimulus or not. Harden et al. (2010), Kosasih and Silver-Thorn (1998) and Quai et al. (2005) looked at pinprick, vibration sense, light touch sensation, deep touch sensation and deep pressure sensation to determine whether amputees have decreased cutaneous sensation on either their amputated leg or the non-amputated one. Most papers looked exclusively at transtibial amputees, except Harden et al. (2010) who also included upper-limb amputees. Quai et al. (2005) looked at exclusively vascular amputees. Both Kosasih and Silver-Thorn (1998) and Harden et al. (2010) had groups of mixed etiology from vascular, traumatic and cancer origins.

Harden et al. (2010) and Kosasih and Silver-Thorn (1998) used a pinprick on an area of skin clear of scar tissue to see if the participant could feel the superficial pain. It was the most affected out of all the tests in all groups of amputees. The amputated leg could not feel the pinprick in all circumstances.

Harden et al. (2010), Kosasih and Silver-Thorn (1998), and Quai et al. (2005) tested vibration sense at the site of bony landmarks on both the amputated and non-amputated legs. This was done with a vibrating fork of 128 Hz and the individual had to determine whether the fork was vibrating or whether it was just the pressure of the non-moving fork that was felt when the fork was pressed to their skin. They found that vibration sense was minimally affected in the amputated leg compared to the non-amputated leg in transtibial amputees.

Harden et al. (2010), Kosasih and Silver-Thorn (1998) and Quai et al. (2005) placed an aesthesiometer on the skin to see if it was felt by the participant to test light touch sensation and deep touch sensation. They all found that light touch sensation was slightly affected on the amputated side compared to the non-amputated side. Quai et al. (2005) explained that light touch sensation is related to which leg the amputees are most likely to bear their weight on. A decrease in sensation was related to more even weight bearing in amputees and decreased light touch sensation meant the amputees bore more weight on the non-amputated leg. The hypothesis was that amputees with better light touch sensation would rely entirely on the non-amputated leg for sensory information, whereas those with a decrease in sensation had to rely on the stump as well for information (Quai et al., 2005). Kosasih and Silver-Thorn (1998) were the only ones to test deep pressure sensation and found it was not affected in amputees compared to controls.

1.3 Proprioception

Proprioception is knowing where a limb is in space without having to look at it or touch it. This is done by receiving information from proprioceptors found in the muscles, ligaments and joints (Hall and Guyton, 2011). The muscle spindle is a good example since it relays information to the central nervous system on the length and velocity of stretch of a specific muscle. Proprioception has been studied in various populations and has been looked at in various joints throughout the body. Seeing as the amputees are missing tissues such as muscles and ligaments at certain joints, and subsequently are missing the necessary receptors, proprioception could be affected just like cutaneous sensation.

There are two main tests used to evaluate proprioception; threshold determination of passive motion (TDPM) and joint repositioning (JR). It has been shown that both transtibial and transfemoral amputees have more trouble with the passive motion detection compared to the JR indicating that the TDPM test is more sensitive in this population (Eakin et al., 1992; Liao and Skinner, 1995). Using the TDPM test, it has been shown that proprioception is affected in the amputated limb when compared to the legs of able-bodied controls. This is also the case when compared to the non-amputated leg. However, there are conflicting results in the non-amputated limb compared to the control legs. These findings do not hold with JR.

1.3.1 Threshold Detection of Passive Motion

When it comes to TDPM, the limb is displaced from a predetermined initial angle. Various studies use different low speeds to move the limb, and the participant must determine the point at which they feel the limb moving, typically by pushing a button to signal that they have felt movement. The outcome measure of interest is the difference in angle from when the machine started to move the leg, until the angle at which the participant perceived movement.

Three studies tested TDPM in lower-limb amputees (Eakin et al., 1992; Kavounoudias et al., 2005; Liao and Skinner, 1995). All three groups tested different joints; with Eakin et al. (1992)

testing the hip and the knee together, Kavounoudias et al. (2005) testing the knee and ankle separately and Liao and Skinner (1995) testing the knee only. Liao and Skinner (1995) and Eakin et al. (1992) studied groups of mixed etiologies mainly composed of traumatic amputees. The study by Liao and Skinner (1995) also included two of vascular origin while Eakin et al. (1992) had one participant with an amputation due to a tumour. Kavounoudias et al. (2005) had separate groups for vascular and traumatic amputees. As for the amputation site, both Liao and Skinner (1995) and Kavounoudias et al. (2005) looked at transtibial amputees, whereas Eakin et al. (1992) looked at transfemoral amputees exclusively. The chosen speeds at which to move the limb ranged from 0.4°/s to 0.7°/s. Eakin et al. (1992) and Liao and Skinner (1995) agreed that the amputated limb has decreased proprioception when compared to the non-amputated limb, whereas Kavounoudias et al. (2005) found that there was no difference in TDPM between the amputated leg and the non-amputated leg in both the traumatic group and vascular group. Both Liao and Skinner (1995), and Kavounoudias et al. (2005) also compared the non-amputated leg to those of an age-matched control group and found conflicting results. Kavounoudias et al. (2005) found that the non-amputated limb had a greater threshold than the controls, indicating decreased proprioception, whereas Liao and Skinner (1995) found that there was no significant difference. Liao and Skinner (1995) was the only study to compare the amputated side to both legs of an age-matched control group and they found that the amputated leg felt the movement at a larger angle.

1.3.2 Joint Repositioning

Another method to measure proprioception is through JR. The limb is placed an initial position and then placed in a specific target position, held for a set time, typically a few seconds, and then placed back to its initial position. The participant is then asked to replicate the target position. This test measures the difference between the target angle and the replicated angle.

Both Eakin et al. (1992) and Liao and Skinner (1995) also looked at JR, along with Latanioti et al. (2013). Latanioti et al. (2013) had a mix of both traumatic and cancerous transfemoral amputees in one group. When comparing the amputated leg to the non-amputated one, they all found that there was no difference in the JR test. Latanioti et al. (2013) hypothesized that the proprioceptive receptors in the hip take over as well as the cutaneous sensation at the stump which explains why there was no difference between the two legs. Liao and Skinner (1995) also found there was no difference between the amputated leg and the non-amputated limb compared to the leg of a control group. Time since amputation was correlated with proprioceptive sense, with amputees that had been living with the prosthesis for longer having better proprioception (Latanioti et al., 2013). This may explain why there was no difference between amputees and controls.

1.4 Strength

Overall, strength is affected in the amputated limb. Of the studies testing strength in amputees, only the knee and hip were assessed. Testing of the non-amputated leg only evaluated strength at the knee and it was shown to be affected. These results do not seem to carry on to the hip of the non-amputated leg.

Amputees are missing muscles which leads to a decrease in muscle mass and an atrophy of the remaining muscles (Fraisse et al., 2008). This decrease in muscle mass could lead to a decrease in strength (Fraisse et al., 2008; Renstrom et al., 1983a). Strength can be measured isometrically and isokinetically, which includes both concentric and eccentric contractions. With isokinetic measurements, the speeds at which a joint is normally tested in amputees range from 30°/s to 180°/s (Isakov et al., 1996b; Moirenfeld et al., 2000; Nolan, 2009; Pedrinelli et al., 2002; Renstrom et al., 1983b; Tugcu et al., 2009). Most studies with amputees analyzed peak torque; other variables considered included average maximal torque, peak bending moment, total work, maximum power and flexor/extensor relation. Several groups tested strength in transtibial amputees, with some having exclusively traumatic groups (Moirenfeld et al., 2000; Tugcu et al., 2009) and most being of mixed nature with traumatic, vascular and cancer all together (Isakov et al., 1996b; Nolan, 2009; Pedrinelli et al., 2002; Renstrom et al., 1983b) whereas type of amputees was not described in a study by Isakov et al. (1996a). Nadollek et al. (2002) was the only study to look at an exclusively vascular group.

All groups reported that the amputated leg is weaker than the non-amputated leg at the knee. Pedrinelli et al. (2002) also compared amputees' non-amputated legs to that of able-bodied controls and found that the amputees' non-amputated leg had a smaller bending moment, decreased work, and less power at the knee. Isakov et al. (1996b) investigated the effect of stump length and time since amputation on strength of the knee muscles. They found that those with a shorter stump were weaker than those with a longer stump (Isakov et al., 1996b). In another study there was no significant difference in knee strength between those whose amputation dates from longer than seven years compared to less than seven years (Isakov et al., 1996a).

There is conflicting evidence regarding the strength at the hip. Nadollek et al. (2002) and Tugcu et al. (2009) found that there is no difference in strength between the amputated leg and the non-amputated leg while Nolan (2009) found the same results in sports-active amputees only. In inactive amputees, there was a decrease in strength in the amputated leg compared to the non-amputated leg.

1.5 Balance

Balance is a complex task that not only involves strength but also the visual system, proprioception and cutaneous sensation. The most common method of measuring balance ability, is to assess the displacement and velocity of COP or to evaluate standing time. Balance is measured under various conditions including quiet standing and perturbed standing either during double or single-legged stance.

Less is known about perturbed balance in amputees than quiet stance. Similar to quiet stance, amputees are less stable than controls during perturbations. The strategies used are also different; where amputees will use more of a hip strategy and able-bodied controls will use more of an ankle strategy.

1.5.1 Balance during Quiet Stance

Balance during quiet standing has been studied in both transfemoral and transtibial amputees under different conditions such as eyes open and closed. Sway, weight bearing distribution and standing time have been assessed in amputees and compared to able-bodied controls. Sway is evaluated through the movement of the COP, including position, net displacement, root mean square (RMS) displacement, and velocity. Sometimes COP movement is separated into two directions; anterior-posterior (AP) and medio-lateral (ML) whereas sometimes the two directions are combined.

Amputation affects balance during both double and single-legged stance and the consensus is that amputees have decreased balance abilities compared to able-bodied controls. COP is the main measure of balance ability, and its motion was found to increase in amputees. The time amputees could remain balanced on one leg before putting a foot down, also known as time in balance, is affected as well, so that in amputees time spent in balance is decreased. It is mostly agreed that amputees bear more weight on their non-amputated leg. This difference in weight bearing may explain why the excursion of the COP under the non-amputated leg is greater than under the amputated leg during double-legged stance. During single-legged stance, balance is decreased on both the amputated and non-amputated leg but only time in balance seems to be affected.

1.5.1.1 Two-Legged Stance: Amputees vs Controls

Quiet standing during double-legged stance has mainly been studied in transtibial amputees (Buckley et al., 2002; Gauthier-Gagnon et al., 1986; Hermodsson et al., 1994; Isakov et al., 1992; Kanade et al., 2008; Vittas et al., 1986). To our knowledge, Buckley et al. (2002) is the only group reporting on both transtibial and transfemoral amputees mixed into one group. As for the type of amputations, Buckley et al. (2002) had a group of traumatic amputees and Hermodsson et al. (1994) had separate traumatic and vascular groups. All other studies included a vascular only group. The general consensus is that amputees sway more than able-bodied controls, regardless of amputee type and testing condition, whether in the AP and ML directions separately or combined into overall sway (Buckley et al., 2002; Hermodsson et al., 1994; Isakov et al., 1992; Kanade et al., 2008). Note that, Kanade et al. (2008) found that amputees swayed more in the AP direction but swayed with the same amplitude in the ML direction compared to able-bodied controls. Hermodsson et al. (1994) came to different conclusions depending on the etiology of amputation. They concluded that vascular amputees swayed more in the ML direction than controls. As for the traumatic amputees, they found that there was less AP sway when compared to both the vascular amputees and the able-bodied controls. Results are conflicting regarding the amount of sway in amputees, since amputees' COP has also been shown to move less than controls (Gauthier-Gagnon et al., 1986; Hermodsson et al., 1994; Vittas et al., 1986). Conflicting results may be due an effect of other variables such as age of participant, type of amputation, time since amputation, stump length, or physical activity level. A study done by Vittas et al. (1986) reported that both older amputees and younger amputees sway less in the AP directions than age-matched controls.

However, only younger amputees also swayed less in the ML directions compared to able-bodied controls. Rehabilitation increases sway in amputees compared to controls and yet amputees still sway less than able-bodied controls (Gauthier-Gagnon et al., 1986). Stump length has also been found to influence the amount of sway, where amputees with a shorter stump have increased sway compared to those with a longer stump (Lenka and Tiberwala, 2010). Hermodsson et al. (1994) also tested time in balance during double-legged stance and found that there was no difference regardless of amputee type compared to controls. Hermodsson et al. (1994) concluded that vascular amputees completed less standing time than controls when eyes were closed. There was no difference in standing time between traumatic amputees and controls during the eyes closed condition whereas the traumatic amputees completed longer standing time then vascular amputees.

1.5.1.2 Two-Legged Stance: Amputated vs Non-Amputated Leg

The balance contribution of the non-amputated and amputated legs have been found to be different in double-legged stance. Hlavackova et al. (2011) had a group of transfemoral amputees exclusively, unlike Nadollek et al. (2002) who had a group of transtibial amputees. Rougier and Bergeau (2009) had separate groups of transtibial and transfemoral amputees. As for the type of amputations, Hlavackova et al. (2011) studied traumatic amputees, while Rougier and Bergeau (2009) and Nadollek et al. (2002) only included non-vascular amputees. Hlavackova et al. (2011) and Nadollek et al. (2002) both reported more sway under the non-amputated leg. Note that Nadollek et al. (2002) only observed more displacement of the COP under the non-amputated limb than the amputated limb in AP. ML displacement on the other hand, was found to be the same for both legs. Rougier and Bergeau (2009) did not compare the non-amputated and amputated legs but compared transfemoral amputees non-amputated leg to that of the transtibial amputees during two-legged standing and detected more movement of the COP in both the AP and ML directions in transfemoral amputees. These same researchers found that the COP of the transtibial amputees, under the amputated leg, was more forward and lateral than that of the transfemoral group. An increase in sway under the non-amputated leg is not surprising since weight bearing is not even between the non-amputated and amputated legs, where more weight is supported by the non-amputated leg (Hlavackova et al., 2011; Isakov et al., 1992; Nadollek et al., 2002; Rougier and

Bergeau, 2009). Increased involvement of the non-amputated leg during double-legged stance could lead to an increase in COP movement.

1.5.1.3 Single-Legged Stance

Single-legged stance in amputees has been studied by three groups; Curtze et al. (2016); Hermodsson et al. (1994); Mayer et al. (2011). All groups were interested in transtibial amputees. As for the etiologies, Mayer et al. (2011) had an exclusively vascular group, Hermodsson et al. (1994) had separate traumatic and vascular groups and Curtze et al. (2016) had a mix of traumatic, vascular, congenital and cancer. The general consensus is that sway is decreased in the amputated leg and is the same in the non-amputated leg compared to controls. However time in balance, the amount of time amputees can remain in balance on one leg, is decreased in both the amputated and non-amputated legs compared to controls. Sway was less under the amputated leg than the non-amputated leg and legs of able-bodied controls during single-legged stance (Curtze et al., 2016). As for the sway on the non-amputated leg, there was no difference when compared to able-bodied controls (Curtze et al., 2016; Hermodsson et al., 1994). Mayer et al. (2011) did not compare the amputees to able-bodied controls, but rather recent amputees to those with an older amputation. In the group with newer amputations, the authors reported decreased sway in both the AP and ML directions compared to those who have lived with their amputations for longer. In the same study, sway was found to move more in the AP direction than in the ML direction in participants with older amputations. However, those with a newer amputation sway equally in the ML and AP direction. Similarly to amputees with less recent amputations, able-bodied participants sway more in the AP direction than in the ML direction during double-legged quiet standing (Blaszczyk et al., 2014).

Time in balance was also measured during single-legged stance. The task in Hermodsson et al. (1994) was to stand on one leg for 30 seconds, which turned out to be too difficult, with 0/36 on the amputated leg and 16/36 on the non-amputated leg able to perform the task. Therefore, the researchers measured standing time and found that vascular and traumatic groups separately and together, stood for less time on their amputated leg than the controls. However there was only a trend in standing time difference between vascular and traumatic amputees with vascular amputees completing slightly less time than the traumatic amputees. As for the one-legged stance time on the non-amputated leg, Hermodsson et al. (1994) concluded that vascular amputees spent less time in balance than traumatic amputees and able-bodied controls separately. Time in balance was not different between the controls and the traumatic amputees when standing on the non-amputated leg. Curtze et al. (2016) also reported that most amputees were not able to stand on the amputated leg for more than two seconds on average.

1.5.2 Perturbed Balance

Perturbations can result from external or internal sources. External perturbations are movements imposed on the individual resulting from the environment around them. Internal perturbations are the result of a movement produced by the individual that requires them to compensate to keep their balance. Sway is affected under both the non-amputated and amputated legs compared to able-bodied control legs. For example, during platform perturbations, COP movement is increased under the non-amputated leg and decreased under the amputated leg, compared to controls. Time in balance is decreased during perturbed standing.

1.5.2.1 Perturbation Platform

Bolger et al. (2014); Molina-Rueda et al. (2016); Nederhand et al. (2012); Vrieling et al. (2008) used perturbation platforms to evaluate balance in amputees. Vrieling et al. (2008) looked at a mix of transtibial and transfemoral amputees whereas Nederhand et al. (2012) had two separate groups of transfemoral and transtibial amputees. Bolger et al. (2014) had exclusively transtibial amputees who were of traumatic origin only. Molina-Rueda et al. (2016) had a group of trauma and a group of vascular transtibial amputees. All of the other studies had mixed etiologies that included trauma, vascular, cancer and congenital. The four research groups used different types of perturbations. Bolger et al. (2014) and Molina-Rueda et al. (2016) set up multi-directional translations, Nederhand et al. (2012) had random continuous movements in all directions and Vrieling et al. (2008) had sinusoidal AP perturbations. In all studies, sway was assessed in both the amputated and non-amputated leg separately when reacting to a perturbation during double-legged stance. Sway in the AP direction in the non-amputated leg was larger compared to controls, whereas sway in the amputated limb was decreased compared to controls (Bolger et al., 2014). Also, the AP ground reaction force under both the amputated and non-amputated legs were larger than controls during double-legged stance with eyes open (Vrieling et al., 2008). When it came to body weight distribution Nederhand et al. (2012) found that even during perturbations, weight bearing was not different between the amputated and non-amputated legs whereas Vrieling et al. (2008) reported more weight on the non-amputated leg than the amputated one. In response to a forward translation of the support surface, Molina-Rueda et al. (2016) report a shorter reaction latency in the non-amputated leg of traumatic amputees compared to vascular amputees and able-bodied control legs. No differences in latencies were found between the vascular group and the controls and the reaction latency was not measured in the amputated leg.

1.5.2.2 Other External Perturbations

Buckley et al. (2002); Curtze et al. (2012); Mohieldin et al. (2010) used different protocols involving external perturbations. Buckley et al. (2002) placed participants on stabilimeters, Curtze et al. (2012) used waist-pull perturbations, Mohieldin et al. (2010) had participants perform the Sensory Organization Test. Buckley et al. (2002) and Mohieldin et al. (2010) looked at a mix of transtibial and transfemoral amputees where Buckley et al. (2002) had exclusively traumatic amputees and Mohieldin et al. (2010) had amputees from vascular and non-vascular causes. Curtze et al. (2012) included exclusively transtibial amputees of mixed etiologies that included trauma, vascular, cancer and congenital. Buckley et al. (2002) tested out either AP or ML stability with eyes open and then ML perturbations with eyes closed using a stabilimeter board and counted more board contacts on the amputated than the non-amputated side for ML perturbations with eyes closed. There was no difference in the number of board touches for the dominant and non-dominant legs of controls. Amputees also spent less time in balance than the controls both in AP and ML perturbations. Curtze et al. (2012) tested waist-pull perturbations and found that when releasing the tension of the pull to induce a fall towards the amputated leg, both hips contributed equally to keeping balance with equal moments, whereas, when inducing a fall towards the non-amputated leg, the ipsilateral hip contributed more to balance recovery with a greater moment produced than the hip on the amputated side. Both these situations are different from the controls; when inducing a fall to the side, it was the contralateral hip that contributed more to balance recovery. Curtze et al. (2012) found that the ankle moment of the amputee’s non-amputated leg is what contributed greatly to keeping their balance. Mohieldin et al. (2010) studied how amputees performed the

Sensory Organization Test compared to controls. The Sensory Organization Test consists of six different conditions where vision, proprioception and vestibular sense are given conflicting information to see how well the participant can react to the different conditions. Overall the amputees had a lower score than the control group, especially for the conditions where at least two out of the three senses were challenged. There was also a significant difference between the vascular and the traumatic amputees where the traumatic group performed better.

1.5.2.3 Internal Perturbations

Both Mouchnino et al. (1998) and Viton et al. (2000) introduced internal perturbations by instructing participants to perform lateral leg raising so that they went from standing on two legs, to balancing on one. Both groups compared traumatic transtibial amputees to controls and were in agreement that the weight transfer phase lasted longer in the amputee group than the control group. They also found that the displacement of the COP was smaller in amputees than controls during the movement initiation for both the amputated and non-amputated sides. According to Mouchnino et al. (1998), the center of gravity was found to move less in imbalanced trials (trials in which the trunk or leg did not stay in the same position for at least 2 seconds) compared to balanced trials for the amputees. In the control group, this analysis was not done since there was an insignificant number of imbalanced trials. During a balanced trial, there was less displacement of the center of gravity in the AP direction for amputees than able-bodied controls. The center of gravity moved the same distance overall in the amputee group regardless of the leg being lifted. This difference in number of balanced and imbalanced trials indicates that lateral leg raising is more difficult for amputees than controls.

1.6 Electromyography

Not only do the supporting muscles need to be strong enough to maintain balance, but they need to fire at the right moment and with the proper intensity. Electromyography (EMG) is used to detect muscle patterns from the most simple of tasks to the more complex multi-jointed movements. The onset of spikes in EMG has been assessed during balance tasks. Whether assessed during internal or external perturbations, the latency of the EMG signals is generally affected by amputation. Different muscles are also recruited in amputees compared to able-bodied controls.

For example, it has been found that amputees recruit more proximal muscles whereas able-bodied controls recruit distal muscles indicating the use of a hip strategy in amputees and an ankle strategy in controls. This difference in recruitment has been shown for both the amputated and non-amputated legs.

Both Rusaw et al. (2013) and Viton et al. (2000) looked at the EMG of transtibial amputees during two different balance scenarios. Rusaw et al. (2013) had a group of mostly traumatic amputees mixed with congenital, infection and cancer whereas Viton et al. (2000) included participants with amputations from traumatic origins exclusively.

Rusaw et al. (2013) examined EMG firing patterns in amputees during double-legged stance on a perturbation platform that tipped to move the toes up or down. They found that muscle activation in both the amputated and the non-amputated legs reacted slower than the legs of the able-bodied controls when the platform tipped. The authors mentioned that since there is a reorganization in the sensory information received from the opposite limb, the EMG response in the non-amputated leg is also delayed. Viton et al. (2000) analyzed balance during standing on both legs to raising one leg laterally, testing both the amputated leg and the non-amputated leg in random order. They found the amputee's medial gastrocnemius, on the non-amputated leg when being raised, fired with the same intensity and duration when compared to the controls. There is no information on the amputated leg since EMG of the medial gastrocnemius was not measured. These same authors also saw that the tibialis anterior fired at the same time as the medial gastrocnemius in the non-amputated legs when they were raised, but this did not occur in the control legs of able-bodied participants. Lastly, they also found that the tensor fasciae latae was the first muscle activated on the leg that was to be raised in amputees. This was true in both the amputated side and the opposite side, but did not occur in the legs of able-bodied controls. However, the burst from the tensor fasciae latae lasted longer when the amputated leg was raised, compared to the non-amputated leg. Viton et al. (2000) suggested that unlike able-bodied controls, amputees use the tensor fasciae latae as the muscle that initiates the medial shift of the COP before lateral leg raising. Instead, in the control legs, the gastrocnemius medialis was the first muscle to fire and initiate the shift in COP. They suggested that since amputees can no longer use the medial gastrocnemius muscle in the amputated limb, proximal muscles such as the tensor fasciae latae compensate to initiate shifting the COP medially when standing on the non-amputated leg. Viton et al. (2000) explained that since amputees are missing proprioceptive information and their cutaneous sensation is diminished, there is a change in the firing pattern of the non-amputated leg to initiate the medial shift of the COP.

1.7 The Relationship between Balance and Other Variables

It has been suggested that balance is affected in amputees and that certain variables such as cutaneous sensation, proprioception and strength are also affected. Few studies have looked at the link between these variables and balance. To our knowledge, a relationship with balance has only been assessed with cutaneous sensation and strength. Both Nadollek et al. (2002) and Quai et al. (2005) tested vascular transtibial amputees. Quai et al. (2005) found that the amputees that did not perform well on the light touch sensation test, experienced greater COP movement in the AP direction under the amputated leg during quiet standing in double-legged stance. They were also at a higher risk of falls. Nadollek et al. (2002) reported an increase in the ML displacement of the COP under the amputated leg, when standing on both legs, in those with weaker hip abductors. Although this correlation did not hold in the non-amputated leg. Those that had stronger hip abductors bore more weight on the amputated leg, swayed less in the ML direction and were less likely to experience pain and took less medications (Nadollek et al., 2002).

CHAPTER 2 − RATIONALE, OBJECTIVES & HYPOTHESES

2.1 Rationale

It has been shown that amputees have decreased balance abilities during double-legged stance. Moreover, there is less sway under the amputated leg than under to non-amputated leg. This could be due to increased weight-bearing under the non-amputated leg. Single-legged stance has been shown to be affected in amputees (Curtze et al., 2016; Hermodsson et al., 1994; Mayer et al., 2011). This is not only true in the amputated leg, as one might expect, but it has been found to also be the case in the non-amputated leg (Hlavackova et al., 2011; Nadollek et al., 2002; Rougier and Bergeau, 2009). Center of pressure (COP) is affected under the amputated leg only but time in balance is affected in both the amputated and non-amputated legs. Lastly, amputation affects the reaction to support surface perturbations in double-legged stance (Bolger et al., 2014; Nederhand et al., 2012; Vrieling et al., 2008). To our knowledge single-legged stance on the non-amputated leg during perturbations has not been studied in amputees. Testing the non-amputated leg only would provide more information on the contribution of that leg to balance and would remove other factors such as the type of prosthetic and mechanical ankle. The majority of studies support the idea that both the amputated and non-amputated legs display decreased cutaneous sensation, proprioception and strength. Modifications of these elements in the non-amputated leg would be expected in vascular amputees with their comorbidities, but it is also the case in traumatic amputees. However, only a few studies have analyzed the relationship between balance and these variables (Nadollek et al., 2002; Quai et al., 2005). Electromyography (EMG) patterns were also shown to be affected during lateral leg raising but were not studied in single-legged stance during perturbations. The overall goal of our study was to assess balance in traumatic transtibial amputees standing on the non-amputated leg during forward translations of the support surface and to assess cutaneous sensation, proprioception and strength and their relationship with balance.

2.2 Objectives

1. Assess balance in lower-limb amputees during single-legged stance on the non-amputated leg during perturbed conditions, by evaluating COP and center of mass (COM). EMG was also measured to assess muscular contribution to balance.
2. Assess cutaneous sensation, proprioception and strength in the non-amputated leg.
   - 2.1 Assess cutaneous sensation using the touch pressure sensation (TPS) test in the non-amputated leg of lower limb amputees.
   - 2.2 Assess proprioception using the threshold determination of passive motion (TDPM) test in the non-amputated limb of lower limb amputees.
   - 2.3 Assess concentric strength at the knee and ankle in the non-amputated leg of lower limb amputees.
3. Study the role of the variables such as proprioception, cutaneous sensation and strength and how they relate to balance as measured through COP and COM variables.

2.3 Hypotheses

1.1 COP excursions will be the same in amputees and in controls before the perturbation; they will increase after the perturbation with a larger increase in amputees than in controls.

1.2 During the perturbation, there will be no differences between amputees and controls in COM and COP excursions.

1.3 EMG patterns will be different in amputees and controls. For example, the amputees may experience a delay in activation.

1.4 The amount of muscle activation will increase after the perturbation, with a larger increase in amputees.

2.1 Cutaneous sensation will be decreased in the non-amputated limb compared to controls.

2.2 Proprioception will be decreased in the non-amputated limb compare to controls.

2.3 Concentric isokinetic contractions will be decreased at the knee and ankle in the non-amputated leg compared to controls.

3.1 TDPM, TPS and strength will be correlated to balance where a decrease in cutaneous sensation, proprioception and strength will indicate decreased balance abilities

CHAPTER 3 − MANUSCRIPT

Traumatic Transtibial Amputees Display Tight Control of Center of Pressure in Response to Perturbations

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3.1 Abstract

In lower-limb amputees, muscles and tissues are missing, potentially affecting balance. Balance in amputees has mainly been studied during quiet standing on both legs. The goal of this study was to assess balance in traumatic transtibial amputees standing on the non-amputated leg during forward translations of the support surface. Each trial lasted 30 seconds, with the perturbation being induced anywhere between the 10th and 20th second. Ground reactions forces, as well as kinematics were recorded throughout the trial. Center of pressure (COP) and center of mass (COM) variables were computed 8s before and 8s following the perturbation. Differences were found in the positions of the COP and COM before and after the perturbation, although there were no differences between groups. COP path root-mean-square (RMSd) also increased after the perturbation while the only difference between groups was a trend in average center of pressure velocity (COPv) as well as a trend in the COPv group*time interaction, displaying an increase in controls after the perturbation. The variability in measures were larger in the control group both before and after the perturbation. This suggests that amputees are restricted during balancing tasks and tightly control sway whereas controls have the choice on whether to restrict sway or not.

Keywords: amputees, balance, center-of-pressure, center-of-mass, perturbation

3.2 Introduction

Amputations affected 1.6 million people in 2005 in the United States, and this number is expected to increase to 3.6 million by 2050, following recent trends (Ziegler-Graham et al., 2008). Vascular disease is the leading cause of amputation followed by trauma due to an accident (Moxey et al., 2010). Traumatic amputations generally affect people under 60 years old, where two-thirds of traumatic amputees are under the age of 45 at the time of amputation, with traffic accidents as the leading cause (Dillingham et al., 1998; Gregory-Dean, 1991; Ziegler-Graham et al., 2008). Fear of falling is a legitimate concern for individuals with amputations, seeing as between 52 and 86% of individuals have fallen the 12 months prior to testing (Kulkarni et al., 1996; Miller et al., 2001a; Miller et al., 2001b). Fear of falling may be due to decreased balance which is potentially affected in lower-limb amputees since muscles and tissues are missing.

Balance is generally affected in various standing conditions in both transtibial and transfemoral amputees, independent of the reason for amputation. Most studies have focused on quiet standing conditions on both legs (Jayakaran et al., 2012; Ku et al., 2014). Amputees regardless of type of amputation and level, have generally been found to sway more than able-bodied controls (Buckley et al., 2002; Hermodsson et al., 1994; Isakov et al., 1992; Kanade et al., 2008). This holds true whether sway is studied in the A-P or M-L directions separately, or combined. However, results are conflicting, as COP has also been shown to move less in amputees during quiet standing (Gauthier-Gagnon et al., 1986; Hermodsson et al., 1994; Vittas et al., 1986). Time since amputation, stump length, amputation level, cause of amputation and rehabilitation are all factors that may explain the conflicting results (Gauthier-Gagnon et al., 1986; Hermodsson et al., 1994; Isakov et al., 1992; Lenka and Tiberwala, 2010; Mayer et al., 2011; Rougier and Bergeau, 2009). While standing on two legs, the non-amputated leg bears more weight which may explain the increase in movement of the COP in the non-amputated leg (Hlavackova et al., 2011; Isakov et al., 1992; Nadollek et al., 2002).

COP is not affected in amputees during single-legged stance while time in balance is. Indeed, COP was found to not be different in amputees of various etiologies when standing on the non-amputated leg, whereas time in balance was decreased compared to able-bodied controls (Curtze et al., 2016; Hermodsson et al., 1994). Time in balance when standing on the non-amputated leg was decreased in the vascular group only, and in the vascular and traumatic groups pooled together, whereas there was no difference between the traumatic group and the control group (Hermodsson et al., 1994). COP was not evaluated since none of the amputees were able to stand 30s on the amputated leg. In the same study, time in balance was also decreased when standing on the amputated leg whether grouping vascular and traumatic amputees together or comparing them separately to able-bodied controls.

COP and COM are also affected under perturbed conditions. Internal perturbations such as lateral leg raising revealed differences in COP and COM movement between amputees and able-bodied controls. The COM position was slightly closer to the prosthetic leg and further behind in amputees before movement initiation; when transferring from double-legged to single-legged stance, the COP and COM moved less in amputees, independent of the leg being lifted (Mouchnino et al., 1998; Viton et al., 2000). When studying perturbed balance in double-legged stance using a perturbation platform, COP is affected under both the amputated and non-amputated leg and is contingent on the type of perturbation (Bolger et al., 2014; Nederhand et al., 2012; Vrieling et al., 2008). During forward perturbations, although COM displacement was not different between groups, COP net displacement was found to be larger in the non-amputated leg compared to both the amputated leg and control legs, but to be less in the amputated leg compared to control legs (Bolger et al., 2014). This may be related to the fact that weight bearing is larger under the non-amputated leg than under the amputated leg during perturbations (Molina-Rueda et al., 2016; Vrieling et al., 2008).

To our knowledge, only three groups have compared single-legged stance in amputees and able-bodied controls (Curtze et al., 2016; Hermodsson et al., 1994; Mayer et al., 2011), and none have looked at single-legged stance during perturbations. The objective of this study was to assess balance in traumatic transtibial amputees before and after perturbed conditions, while standing on the non-amputated leg. We hypothesized that COP and COM excursions would be similar before, but larger after the perturbation in amputees than controls indicating less balance abilities.

3.3 Methods

3.3.1 Participants

We received 61 names of former patients with a lower-limb amputation from a rehabilitation institute through the archives department; of those, 22 did not fit the exclusion/inclusion criteria and 29 refused to participate. Out of the 10 individuals who accepted to participate, four could not stand on their non-amputated leg for at least 30s. In the end, six individuals with a unilateral and transtibial traumatic amputation at least one year prior participated in this study (4M/2F, age: 40.7±10.0 yrs, height: 1.7±0.1 m, body mass: 83.4±23.1 kg, BMI: 28.10±8.71 kg/m^2, time since amputation: 9.8±8.8 years). In two of the amputees, the amputation was due to a work accident and in the four others the amputation was due to a motor-vehicle accident. The control group was matched for age, sex and physical activity level. Six able-bodied controls participated in this study (4M/2F, age: 39.2±11.0 yrs, height: 1.7±0.1 m, body mass: 74.6±13.5 kg, BMI: 24.33±1.65 kg/m^2). All participants did not have a condition that could affect balance such as; a vestibular problem, decreased non-corrected vision, or any past or present neurologic or musculoskeletal conditions, or a condition affecting cutaneous sensation such as diabetes. They all read and signed a consent form outlining the protocol and approved both by the

Concordia University and the Center for Interdisciplinary Research in Rehabilitation of Greater Montreal ethics committees.

3.3.2 Physical Activity Level Questionnaires

The Tegner Activity Level scale and the Human Activity Profile (HAP) questionnaire were used to assess activity level in both activities of daily living and leisure activities. The Tegner Activity Level scale includes daily activities, work-related physical activity and leisure activity evaluated on a scale of 0 to 10 and has been shown to be reliable and valid in individuals with ACL ligament injuries (Briggs et al., 2009; Tegner and Lysholm, 1985). The lowest activity level score is a zero which corresponds to someone on sick leave from work. The highest activity level score is a ten and corresponds to someone who plays sports at a national competitive level. The HAP measures daily activities and their relative energy expenditure and has been tested as reliable and valid in various diseased populations (Davidson and de Morton, 2006). It consists of 94 questions ranging from; getting in and out of chairs without assistance, to running or jogging 3 miles in 30 minutes or less. There are three options for each question; still doing this activity, have stopped doing this activity, and never did this activity. Two values are calculated; the Maximum Activity Score (MAS) is the number representing the most difficult activity the participant can still do and the Adjusted Activity Score (AAS) is calculated by subtracting from the MAS the number of activities below the MAS that the participant stopped doing.

3.3.3 Perturbations

Perturbations were induced using a custom-made platform (H2W Technologies, Santa Clarita, CA, USA, see appendix B) while participants stood quietly on one leg. Amputees stood on their non-amputated leg while controls stood on a self-selected leg, which is the leg they spontaneously transferred to when asked to stand on one leg. Participants placed their hands on their hips, looking straight ahead at a marker at eye level fixed to a wall approximately 3.5 m away. Ten 30-s trials were recorded. In five randomly selected trials, a forward translation was induced anytime between the 10th and 20th second (distance = 9.0 cm, max velocity = 38.0 cm/s, max acceleration = 285.7cm/s^2, total time = 500ms). Approximately 30 seconds elapsed while setting up the system for the next trial, during which participants stood on both legs and placed more weight on the non-tested leg. After each block of five trials, participants sat down for three minutes. Additional breaks were allowed if needed. Lastly, if the contralateral foot touched the platform or if balance was lost, the trial was deemed unsuccessful and was discarded. Additional trials were performed until five successful perturbed trials were recorded, up to a maximum of 15 trials.

3.3.4 Data Collection

During the perturbations participants stood on a force plate (AMTI, Watertown, MA, USA) bolted to the perturbation platform. Data from the force plate was recorded (sampling frequency = 1500Hz) through the Nexus 1.8.5 software (Vicon Motion Systems Ltd, UK) which used the information to compute COP. Kinematics were also recorded using the Vicon motion detection system (Vicon Motion Systems Ltd, UK) with eight MXT20 cameras (sampling frequency = 100Hz). Force plate and kinematics data were synchronized through the Nexus software. Markers were placed on anatomical landmarks according to the Plug-in-Gait marker set (Vicon Motion Systems, 2010). The participant's global COM position was computed through the Nexus software using anthropometric measurements combined with markers' trajectories. Markers were also placed on the four corners of the perturbation platform to follow its displacement. A harness attached to the ceiling was used to prevent falls.

3.3.5 Procedures

Participants arrived in the lab and experimental procedures were explained. Informed consent was signed, and the Tegner activity level scale and the HAP were filled out. To prepare for the balance testing, anthropometric measurements were taken and markers were placed according to guidelines. Skin was prepared and surface electrodes were placed on the rectus abdominis, rectus femoris, vastus medialis, tibialis anterior, peroneus longus, erector spinae, biceps femoris and gastrocnemius medialis according to SENIAM guidelines (SENIAM website). Muscle testing was then performed to ensure proper signal. The results of the electromyography recordings are not reported in this paper. Balance testing was run where five out of ten trials were randomly perturbed. If necessary, additional trials were added with a maximum of 15 trials.

3.3.6 Data Analysis

A 6 Hz zero-lag 4^{th} order low-pass Butterworth filter was used to clean marker and COM trajectories as well as COP data. A 6Hz filter has been previously used by Termoz et al. (2004) to clean COP data. COM and COP data was normalized to the heel position. The timing of the start and end of the perturbation was computed from the movement of the average position of the four markers on the platform, where 5% of the maximum velocity was taken as the threshold. COP and COM average position, COPv, RMSd, and COP velocity RMS (RMSv) were analyzed in the anterior-posterior (AP) axis over 8 seconds both preceding (-8s to onset) and following the perturbation (2s to 10s post onset). Ratios (post/pre) were calculated for the RMSd, COP velocity and RMSv data in order to determine whether post values were smaller or larger than pre values. All analyses were performed using Matlab R2015a (MathWorks, Natick, MA,USA).

3.3.7 Statistical Analysis

T-tests were run on the Tegner and HAP (MAS & AAS) data to compare groups. COM and COP variables were averaged across trials for each participant. Repeated measures ANOVAs were run on the COM and COP average positions, RMSd, COPv, and RMSv with one between (group) and one within (pre/post) factor. A t-test was run on the ratios of the RMSd, RMSv and COPv data. After noticing a large variance in the control group, Levene's test of homogeneity of variance was run to compare the group variance for each of the RMSd, COPv, RMSv variables and ratios. $p<0.05$ was considered significant. All statistical analyses were performed using SPSS version 22 (IBM, Armonk, NY, USA).

3.4 Results

3.4.1 Center of Pressure & Center of Mass Position

Scores for the Tegner Activity Level Scale were similar for the amputees and the controls (amputees: 3.7±2.1, controls: 4±1.1; $t_{10}=0.349$, $p=0.734$). The MAS and AAS scores of the HAP differed between both groups with the amputees having lower scores (MAS: amputees: 77±16.7, controls: 92.7±2.2; $t_{5.170}=2.540$, $p=0.050$, AAS: amputees: 68.7±20.6, controls: 90.8±4.7; $t_{5.510}=2.567$, $p=0.046$).

The average COM and COP curves in AP, relative to the heel, throughout time for each group as well as five individual trials in one participant are presented in figure 3.1. The initial positions for both COM and COP were around 140mm in front of the heel. In response to the perturbation, both COM and COP moved back, with the COP moving further backwards to keep the COM within the base of support. Coming back after the perturbation, the COM of both groups moved anteriorly, where the amputees kept their COM further back (closer to the heel) than the controls. COP position overshot in some controls whereas the amputees came back more gradually. Average COM and COP positions after the perturbation were closer to the heel for both groups (Figure 3.2; COM: $F_{1,10}$=27.958, p<0.001 ; COP: $F_{1,10}$=38.595, p<0.001). There was also a trend in the group*time interactions for COM and COP average position with a larger displacement towards the heel in the amputees than the controls after the perturbation (COM: $F_{1,10}$=4.801, p=0.053, power=0.508; COP: $F_{1,10}$=3.739, p=0.082, power=0.416).

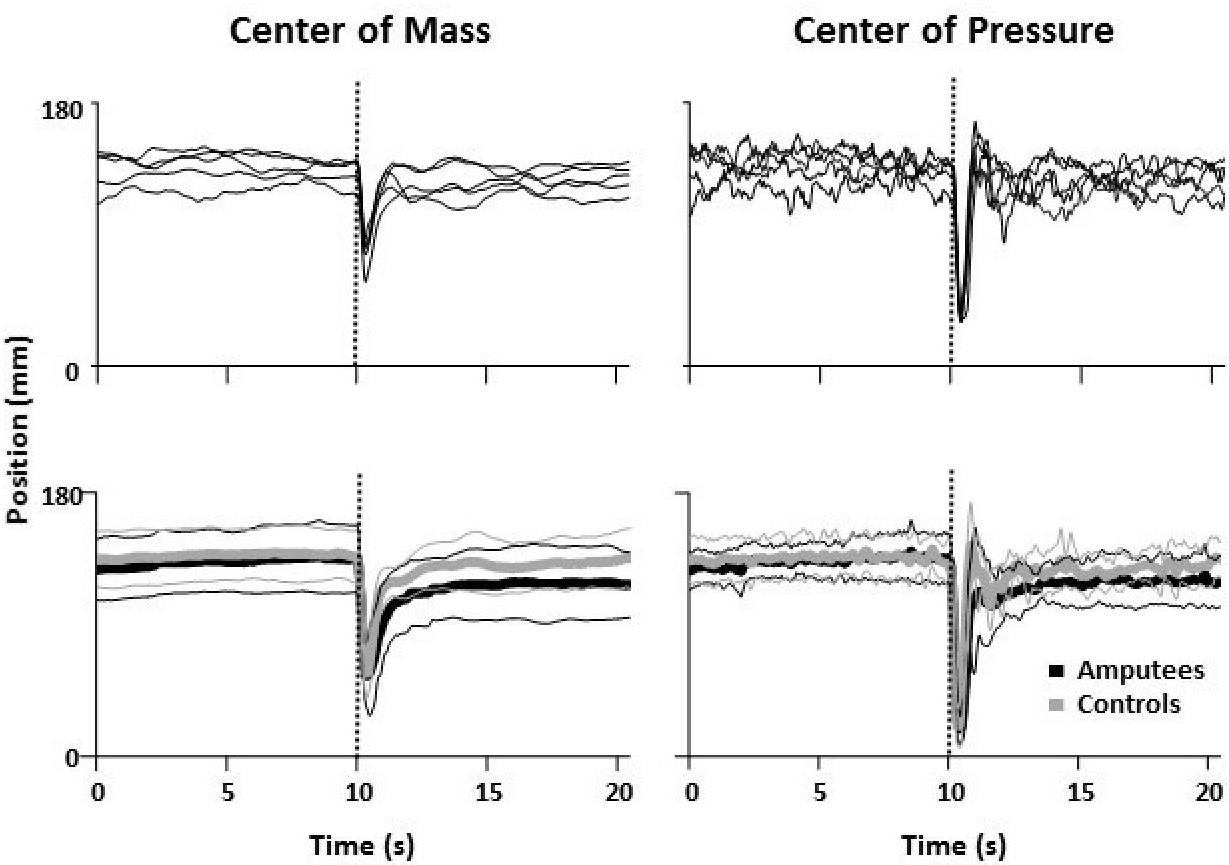

Figure 3.1. Mean center of mass (COM) and center of pressure (COP) position in the anterior-posterior (AP) axis relative to the heel through time. The top panels display all five perturbed trials for one participant, where each line represents one trial. The bottom panels display the average of each group. Thick lines represent group means and thin lines represent one standard deviation above and below the mean. All trials were aligned with the perturbation onset fixed at 10s. The vertical dotted line represents the onset of the perturbation.

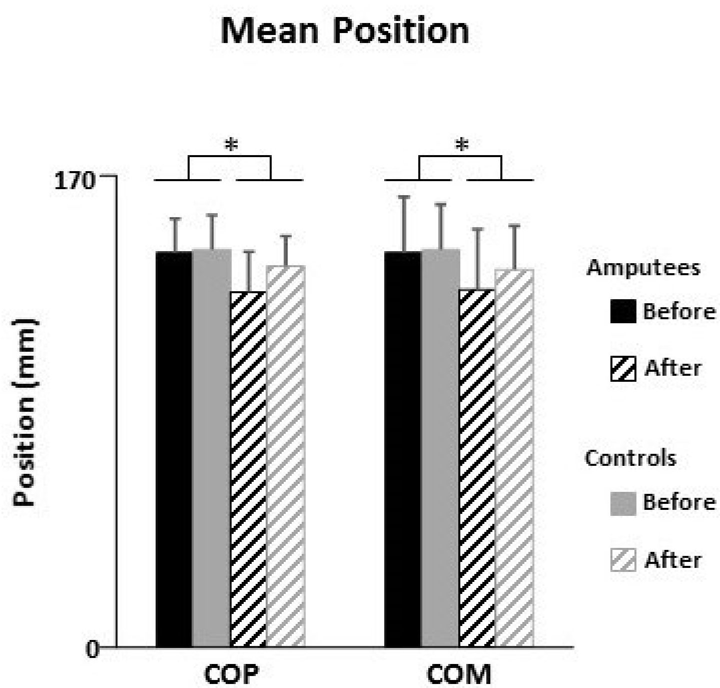

*Figure 3.2. Mean COM and COP position in the AP axis, relative to the heel, before and after the perturbation in each group * p < 0.05*

3.4.2 Center of Pressure Path Root-Mean-Square, Velocity Root-Mean-Square and Average Velocity

RMSd was the only COP variable to be affected by the perturbation with larger values following the perturbation (Figure 3.3: $F_{1,10}$=7.335, p=0.022). There was no difference between groups, except for a trend for greater COPv in controls ($F_{1,10}$=3.660, p=0.085, power=0.409). Group*time interaction for COPv also revealed a trend ($F_{1,10}$=3.173, p=0.105, power=0.364), with a larger increase in controls following the perturbation.

The amputees as a group tended to react similarly and were all grouped towards the lower end, whereas the controls overlapped with the amputees but also spread higher up (Figure 3.4). Amputees were less variable than the controls in RMSd, COPv and RMSv, both before and after the perturbation. More specifically, amputees showed less variability than controls in COPv and RMSv before and after the perturbation, while RMSd showed less variability before the

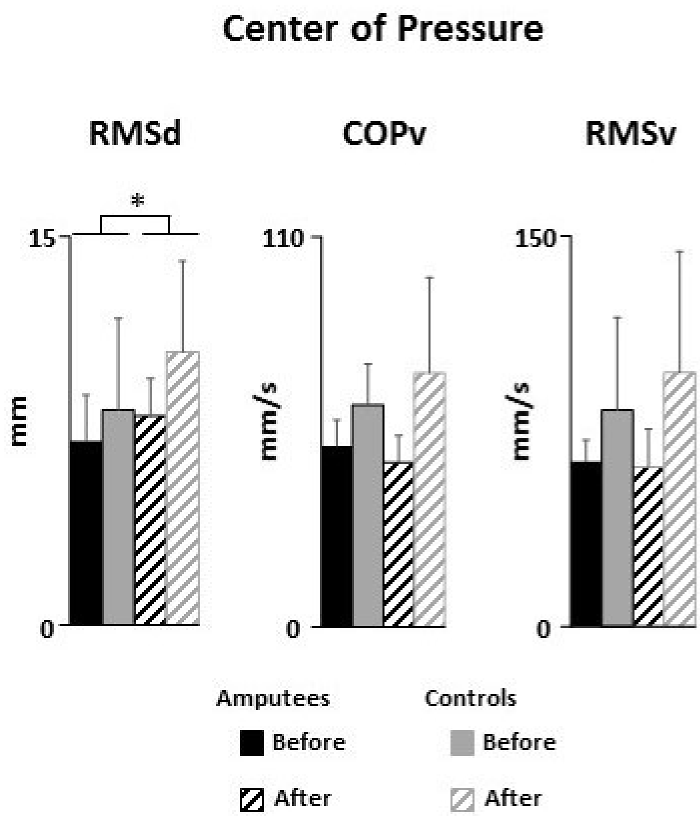

*Figure 3.3. Mean path root-mean-square (RMSd), velocity of the COP (COPv) and velocity RMS (RMSv) in the AP axis before and after the perturbation in each group. * p < 0.05*

perturbation only (COPv: before: $F_{1,10}$=13.297, p=0.004, after: $F_{1,10}$=8.946, p=0.014; RMSv: before: $F_{1,10}$=11.124, p=0.008, after: $F_{1,10}$=7.887, p=0.019; RMSd: $F_{1,10}$=5.078, p=0.048).

Ratios were calculated for RMSd, COPv, and RMSv, where the value after the perturbation was divided by the value before the perturbation (Figure 3.5). A ratio above 100% indicates a larger value while a ratio under 100% indicates a smaller value after the perturbation. The ratio for RMSd was above 100% for both groups indicating an increase in RMSd after the perturbation. However COPv and RMSv were above 100% only in the control group, while they were below 100% in the amputee group. Only the COPv ratio revealed a trend, for a larger ratio in controls

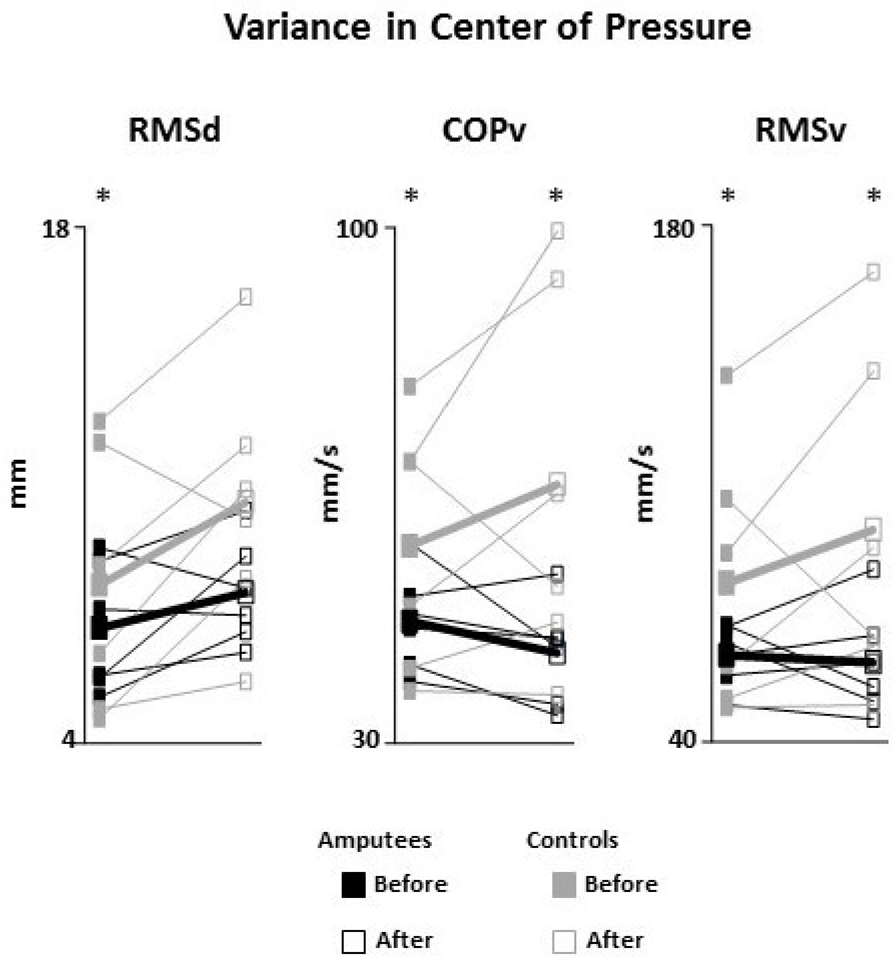

*Figure 3.4. Variability in RMSd, COPv and RMSv in the AP axis, before and after the perturbation in each group. * p < 0.05*

(t_{10}=2.093, p=0.063). Levene's test of homogeneity of variance was not significant for the RMSd, COPv, RMSv ratios.

3.5 Discussion

The goal of the study was to evaluate single-legged stance in traumatic amputees before and after a forward translation of the support surface. The COM and COP did not return to the same position after the perturbation but there was no difference in position between groups.

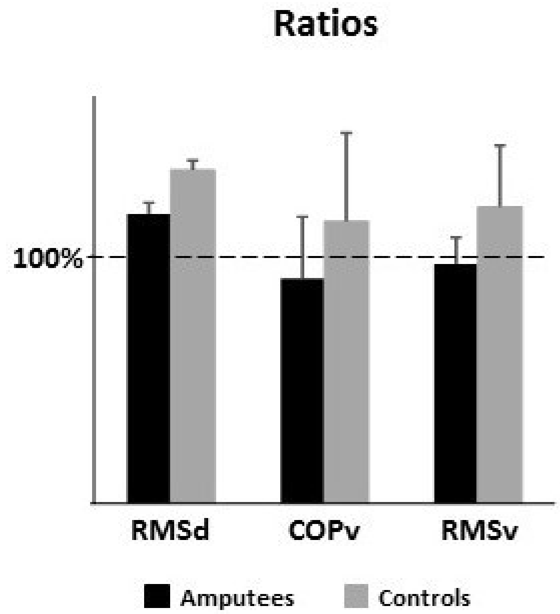

Figure 3.5. Mean post/pre ratios of the perturbation for RMSd, COPv and RMSv in the AP axis in each group.

Overall there was more sway in both groups after the perturbation as portrayed by larger RMSd values. However, sway was not affected by the amputation, displaying similar values for both groups, except for a trend in COPv being larger in controls. The variability in RMSd, COPv and RMSv was larger in the control group, with all the amputees at the lower range of the COP values, and the controls spreading over a larger range of values, both before and after the perturbation.

3.5.1 Center of Pressure & Center of Mass Position

In our study, the COM of both groups was positioned approximately 10cm in front of the ankle before the perturbation. This demonstrates that the participants were already in a more forward position while waiting for the perturbation, as in quiet double-legged standing, the COP is typically positioned approximately 4-6 cm in front of the ankle joint (Danis et al., 1998; Opila et al., 1988). In response to the forward translation of the support surface, we observed both the COM and COP moving towards the heel, with the COP moving further back to keep the COM within the base of support (Winter, 1995; Winter et al., 1998). Unlike the controls, the amputees' COP did not overshoot before stabilizing in the final position. From visual inspection of a figure in a study by Côté et al, it can be seen that the whiplash-associated disorder group's COM overshot the final position following the forward translation of the support surface (Côté et al., 2009). The perturbation was applied with the participants in a seated position where the base of support is much larger than when standing on one foot. Since the sitting position is a safer situation, the whiplash-associated disorder group could adopt a floppy attitude rather than strictly controlling their posture similar to the amputees in our study who may not feel as comfortable standing on one leg during a perturbation thereby exhibiting a stiffer response. The COM and COP eventually returned towards the toes when regaining balance following the perturbation but remained closer to the heel as compared to before the perturbation, more specifically 8cm and 9cm in front of the ankle in amputees and controls respectively, shifting closer to a more typical position. However, there was no difference between groups both in the initial and final position. In the same study by Côté et al., again through visual inspection of the figure, participants did not return to the same initial position after the perturbation instead remaining in a more forward position.

3.5.2 Motion of the Center of Pressure Before and After the Perturbation

In our study, RMSd was larger after the perturbation while recovering balance. Similarly, sway has been shown to increase during single-legged stance compared to double-legged stance, since standing on one leg is a more difficult task (Mayer et al., 2011). In our study, standing after a perturbation may be a more difficult task than before the perturbation, therefore resulting in an increased amount of sway. However, sway did not differ between the amputees and controls both before and after the perturbation, except for a trend in COPv. This is in agreement with previous literature reporting the same amount of COP path and RMSd during quiet standing in amputees standing on the non-amputated leg and controls standing on one leg (Curtze et al., 2016; Hermodsson et al., 1994). In Hermodsson et al.'s study, not all amputees could stand on the non-amputated leg for 30s. During recruitment for our study, four amputees were also not able to stand on the non-amputated leg for 30s and were therefore not included in the testing procedures. Although 16/36 amputees could perform the full 30s trials on the non-amputated leg in Hermodsson et al.'s study, none of them could stand on the amputated leg for 30s. This may explain why more weight is placed on the non-amputated leg during double-legged stance (Hlavackova et al., 2011; Isakov et al., 1992; Nadollek et al., 2002). Amputees thus seem to rely more on the non-amputated leg to compensate for the amputated leg during double-legged stance. However they seem not to compensate enough which results in more sway in amputees than in able-bodied controls (Buckley et al., 2002; Hermodsson et al., 1994; Isakov et al., 1992; Kanade et al., 2008). The increased sway could also result from uneven weight bearing, which has been found to increase sway in able-bodied participants (Anker et al., 2008). Note however that amputees have also been found to sway less than controls in quiet double-legged stance (Gauthier-Gagnon et al., 1986; Hermodsson et al., 1994; Vittas et al., 1986). Conflicting results may be due to time since amputation, rehabilitation, stump length, amputation level and cause of amputation. Indeed those variables have been reported to affect sway in amputees (Gauthier-Gagnon et al., 1986; Hermodsson et al., 1994; Lenka and Tiberwala, 2010; Mayer et al., 2011; Rougier and Bergeau, 2009). For example, increased time since amputation and having a longer stump has been shown to decrease sway while rehabilitation has been shown to increase sway although remaining less than in able-bodied controls (Gauthier-Gagnon et al., 1986; Lenka and Tiberwala, 2010; Mayer et al., 2011). Moreover, transfemoral amputees sway more than transtibial amputees, and vascular amputees sway more than amputees of traumatic origin (Hermodsson et al., 1994; Rougier and Bergeau, 2009).

3.5.3 Floppy vs Stiff Response

Although we did not observe a difference in sway between groups in the current study, variance in COP variables was larger in controls, with amputees displaying lower values and controls displaying a larger range of values overlapping with the amputee group and spreading higher up. To our knowledge, variability in balance control has not been reported, although error bars in a figure comparing sway in transtibial vascular amputees and controls during double-legged stance demonstrates more variability in the control group, with error bars of the control group three times the size the error bars of the amputee group (Gauthier-Gagnon et al., 1986). Amputees may chose to remain stiffer and restrict their sway because they lack balance confidence. In fact, amputees have been reported to not be confident in their balance abilities (Mandel et al., 2016; Miller and Deathe, 2011; Miller et al., 2001a). This is further explained by the fear of falling, which is a common problem in the amputee population where 49.2% of them report such a fear (Miller et al., 2001b). This is a legitimate concern seeing as between 50 and 60% of amputees report falling in the last year (Kulkarni et al., 1996; Miller et al., 2001a). On the other hand, controls can choose between a floppy or stiff configuration allowing themselves to sway more or less since balance confidence is not a factor. Similar to amputees, patients with Parkinson's disease have decreased balance abilities (Blaszczyk et al., 2007) and during forward translations of the support surface, patients with Parkinson's disease have increased muscular co-contraction and in turn, decreased sway, compared to able-bodied controls (Horak et al., 1992). In our study, the amputees may have co-contracted leading to a stiffer response and decreased sway following the perturbation. In fact, amputees co-contract more than able-bodied controls during gait (Seyedali et al., 2012). It is also possible that amputees co-contract more than able-bodied controls during a perturbation or in any situation where their balance is compromised.

In our study, the post/pre ratio for COPv displayed a trend to be larger in able-bodied controls, further demonstrated by the time*condition interaction with the controls increasing the COPv after the perturbation and the amputees decreasing slightly. This relates to the difference in response to the perturbation demonstrated by both groups. Some of the controls use a more floppy response, therefore increase their COPv after the perturbation, whereas the amputees have a more stiff response and do not increase their amount of sway.

3.6 Conclusions

Balance standing on the non-amputated leg is rarely evaluated and to our knowledge, has not been measured during perturbed stance. The amount of sway did not differ between amputees' non-amputated leg and that of able-bodied controls although there was a trend in the COPv interaction with sway in controls increasing more after the perturbation. However, controls displayed more variability in their response. Amputees seem to adopt a more stiff response to a perturbation whereas controls use a variety of responses, ranging from stiff to floppy. This could be linked to decreased balance confidence in amputees.

Acknowledgements

This work was supported by Concordia University PERFORM Center seed funding.

CHAPTER 4 − ADDITIONAL MEASUREMENTS

Additional measurements were collected that were not reported in the article. These include: cutaneous sensation, proprioception, strength and electromyography in the non-amputated leg. These measurements were compared between and within groups and were also correlated to the center of pressure (COP) and center of mass (COM) measures from the balance protocol. More analyses were also performed on the COP and COM data during the perturbation. In this chapter the additional measurements and analyses are reported and discussed. The methods and results presented in the article will not be repeated in this chapter.

4.1 COP, COM, COP-COM: Minimum, Maximum, Onset, Offset

4.1.1 Methods

4.1.1.1 Data Analysis

Data collection is mentioned in the article, so only additional analyses are presented. The onsets and offsets were calculated using 5% of the curve's maximum velocity as a threshold. The timing of all variables is reported relative to the platform onset. The timing of the onset of the platform translation was computed from the movement of the four markers on the platform, where 5% of the max velocity was taken as the threshold. The difference in COP and COM movement was calculated through time for each trial. COP-COM is proportional to the acceleration of the COM, where a large value of COP-COM indicates a large acceleration of the COM (Winter, 1995; Winter et al., 1998).

For the COP, COM and COP-COM variables, the onset, offset, time-to-peak and peak amplitudes were found in the anterior-posterior (AP) axis only. COP, COM and COP-COM all exhibited minimum peaks in their curves whereas COP-COM was the only variable with clear maximum peaks.

4.1.1.2 Statistical Analysis

COM, COP and COP-COM variables were averaged across trials for each participant. T-tests were run on the timing and amplitude variables for COP, COM and COP-COM comparing the two groups. $p<0.05$ was considered significant.

4.1.2 Results

The average COP-COM curve in the AP axis throughout time for each group is presented in figure 4.1. COP-COM oscillates around zero before and after the perturbation where the COP has tight control over the COM. After the perturbation occurs, the COP-COM moves in the negative direction indicating that the COP moves backwards to keep the COM within the base of support. Following the negative peak, a positive peak is seen stabilizing the COM in its final position. The peaks are indicative of a large difference in position between the COP and COM.

Figure 4.2 displays the latency of the onsets, minimum, maximum and offsets for COP, COM and COP-COM. The onset as well as the time to minimum occurred earlier in the COM than in the COP indicating that the COP moves to correct the position of the COM. A trend was revealed for a difference in COP-COM minimum latency with the amputees reacting faster than the controls (t_{10}=1.804, p=0.101). All other latencies to onset, maximum, minimum and offset for COP, COM

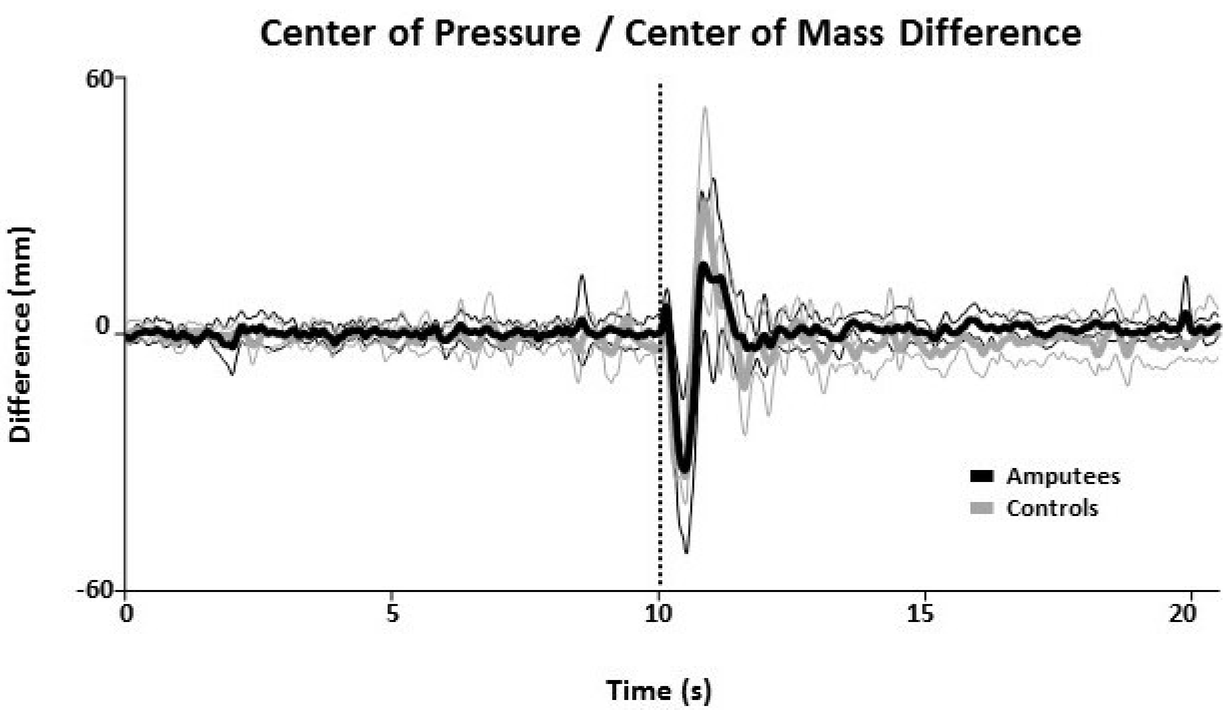

Figure 4.1. Mean difference between center of pressure (COP) and center of mass (COM) position in the anterior-posterior (AP) axis through time in each group. Thick lines display group means and thin lines display one standard deviation above and below the mean. All trials were aligned with the perturbation onset fixed at 10s. The vertical dotted line represents the onset of the perturbation.

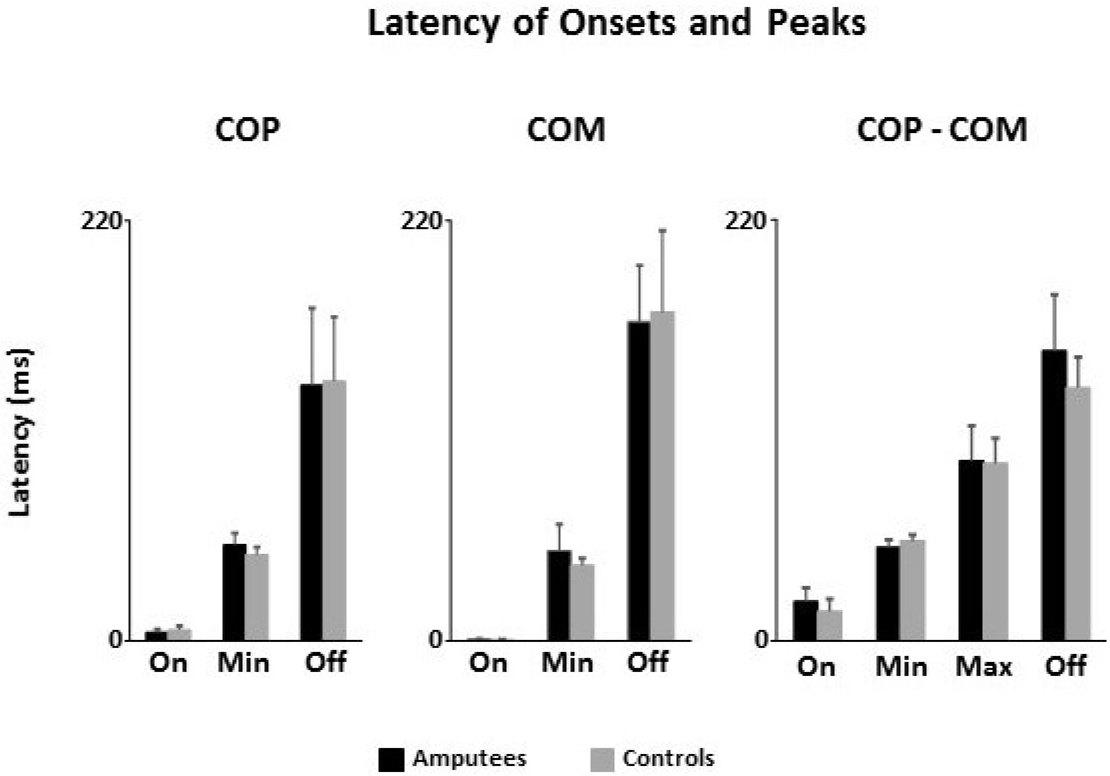

Figure 4.2. Mean latencies for COP, COM and COP-COM of onsets, minimums, maximums and offsets in the AP axis in each group.

and COP-COM were not significantly different between groups. Peak values in COP, COM and COP-COM did not differ between groups (Figure 4.3).

4.1.3 Discussion

The goal of this study was to assess balance in traumatic transtibial amputees standing on the non-amputated leg during forward translations of the support surface. In this section, the timing and amplitude of reactions will be discussed. Minimum and maximum amplitudes for COP, COM and COP-COM were not different between groups. Our study is in agreement with the literature, with amputee's peak COM and COP displacement not different from controls during a forward perturbation while in double-legged stance (Bolger et al., 2014). In our study there were also no differences between groups for most of the latencies of the onset, minimum, maximum and offset for the COP, COM and COP-COM variables except for a trend for amputees to reach the minimum

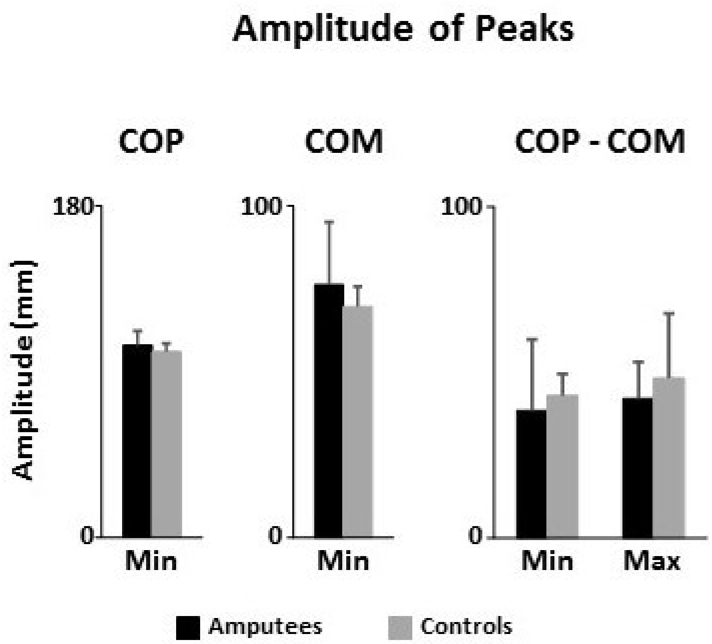

Figure 4.3. Mean amplitude for COP, COM and COP-COM minimums and maximums in the AP axis in each group.

for COP-COM first. Amputees' faster reaction time may be due to decreased balance confidence since they are more concerned with falling (Mandel et al., 2016; Miller and Deathe, 2011; Miller et al., 2001a). Amputees reacted faster but did not react more than controls, seeing as the peak amplitude of COP-COM was the same for both groups. The latency of reaction is similar to Molina-Rueda et al. (2016) where they reported that traumatic amputees had faster reaction times following anterior support surface translations compared to vascular amputees and able-bodied controls.

4.2 Cutaneous Sensation, Proprioception & Strength

4.2.1 Methods

4.2.1.1 Data Collection & Procedures

Three additional measurements were recorded on the same leg as the balance testing; cutaneous sensation, proprioception and strength. The Semmes-Weinstein monofilament test is valid and reliable to assess cutaneous sensation on healthy feet if performed by the same evaluator (Collins et al., 2010; Jerosch-Herold, 2005). TDPM testing has been demonstrated to be reliable and valid under weight-bearing conditions when testing proprioception at the ankle (Deshpande et al., 2003) and non-weight bearing conditions at the knee (Nagai et al., 2012). Isokinetic strength testing at the knee and ankle using a dynamometer have been shown to be valid and reliable (Gaines and Talbot, 1999).

Participants signed the consent form and filled out the physical activity level questionnaires. TPS was then evaluated, first on the plantar site and then on the tibial site. Participants were weighed for the TDPM test by placing both feet on the platform with the load cell. The moveable platform moved six times in extension and six times in flexion in a random order. The balance test was performed third as explained in the article. Lastly, strength testing was done concentrically at the knee and then the ankle in both directions.

TPS was evaluated on a tibial (10 cm below the tibial tuberosity) and plantar site (at the third metatarsal head) using a 20-piece Semmes-Weinstein monofilament kit (North Coast Medical Inc, Gilroy, CA, USA) ranging from 0.008g to 300g of pressure. Although the monofilament was always applied to the skin, participants were told that the monofilament may or may not touch their skin. The participant was asked to avert their eyes from the testing procedure while the monofilament was applied on the skin at 90° with enough pressure for it to arc. The monofilament was briefly held on the skin as the experimenter said: 'Now?'. Participants had been instructed to quickly answer 'yes' or 'no' without hesitation. The first part of the test consisted of finding the estimate threshold monofilament and was done once for each site tested. In order to estimate the threshold, monofilaments were applied starting with the smallest and moving up the set one by one, until participants signaled the first monofilament to be felt. The monofilament was retained as the estimate threshold monofilament and was kept together with three monofilaments above and below, for a total of seven, for the second part of the test which was performed three times, always using the same set of seven monofilaments, and consisted of using the staircase method. Starting with the smallest of the seven selected monofilaments and going in ascending order, the monofilaments were pressed on the participant's skin until they indicated feeling the monofilament, which was kept as the ascending value. Starting with the monofilament that was first felt on the way up, monofilaments were pressed in descending order until the participant did not feel a monofilament, which corresponded to the descending value. The staircase method was performed three times, resulting in three ascending values and three descending values.

Proprioception was evaluated using the TDPM test under weight-bearing conditions. The custom set-up included a moveable platform and a fixed platform (see appendix B). Participants stood with the leg to be tested on the moving platform, while the other leg was on the fixed platform. The initial position of the moving platform was higher so the knee was flexed at 20°. Arms were crossed on the chest. A load cell under the fixed platform was used to ensure the weight was evenly distributed between both legs, within a range of ±10%. The platform moved at a speed of 0.7°/s, either upwards to flex or downwards to extend the lower limb joints. Starting from the initial 20° angle at the knee, the platform moved six times in flexion and six times in extension, in a random order. Participants pressed a button to stop the platform to indicate the moment they felt the motion of the joints. To eliminate auditory cues from the motors, and visual cues, we played white noise through headphones and blocked the view from the legs.

Strength was tested at the knee and ankle using a Humac Norm dynamometer (CSMI, Stoughton, MA., USA). Participants were placed into position according to the manufacturer's guidelines for the muscle being tested and were strapped in to reduce extraneous motion (see appendix C). They moved their limb to their extreme angular position in both directions and five degrees were removed on either end to set the testing range of motion. For each joint, ten cycles of isokinetic concentric contractions in both flexion and extension were performed at a speed of 100°/s as a warm up. After a one minute rest period, six cycles of concentric contractions were performed at a speed of 180°/s. Participants were instructed to contract as forcefully as possible during the procedure and verbal encouragement was given.

4.2.1.2 Data Analysis

TPS values were computed from the average of the three ascending and the three descending values for both the tibial and plantar site. TDPM was estimated from the angular displacement from the initial 20°, to when the button was pushed to indicate feeling the movement. The mean of the six trials in extension and the mean of the six trials in flexion were calculated. Lastly, peak torque over body weight was outputted for both flexion and extension for the knee and dorsiflexion and plantarflexion at the ankle. A flexors/extensors ratio was calculated at the knee, and dorsiflexors/plantarflexors ratio at the ankle.

4.2.1.3 Statistical Analysis

2X2 ANOVAs were run on the TPS and TDPM data with one within (location or direction) and one between (group) factor. A 2X2X2 ANOVA was used for the strength data with two within factors (joint and direction) and one between (group) factor. The strength ratios at the ankle and knee were analyzed with a 2X2 ANOVA with joint as the within factor and group as the between factor. Correlations were also run between the balance variables (Before and after the perturbation: COM and COP mean position, COP path root-mean-square (RMSd), average velocity (COPv), velocity RMS (RMSv); COP, COM, COP-COM: latencies and amplitudes) and cutaneous sensation, proprioception and strength both for the amputees and controls separately and pooled together. $p<0.05$ was considered significant.

4.2.2 Results

As seen in figure 4.4, no significant differences were found in the TPS test between testing sites or between groups, nor was there an interaction.

For the TDPM test, there was a trend for the angle at which a participant felt the movement of the platform to be larger in flexion than extension (Figure 4.5; $F_{1,9}=3.820$, $p=0.082$). Group results for the TDPM test were not significantly different.

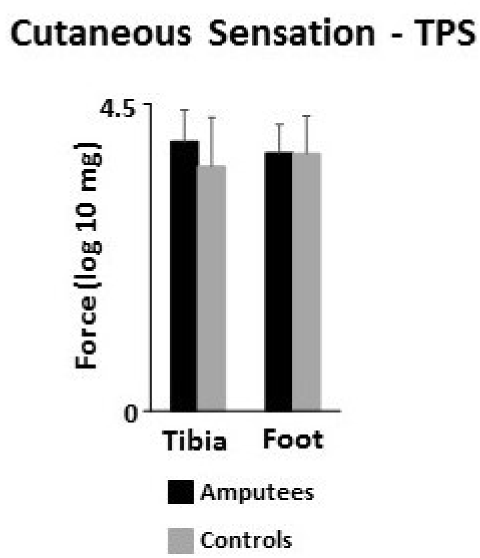

Figure 4.4. Mean touch pressure sensation (TPS) values for the tibial and plantar sites in each group.

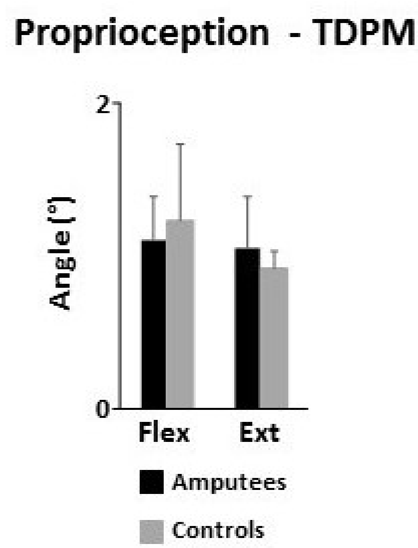

Figure 4.5. Mean angle for threshold detection of passive motion (TDPM) in flexion and extension in each group.

Strength testing revealed a trend for the controls to be stronger than the amputees (Figure 4.6; $F_{1,7}$=5.240, p=0.056). Joints as well as direction were significantly different (joints: $F_{1,7}$=85.085, p<0.001; direction: $F_{1,7}$=308.305, p<0.001). Strength ratios at the knee (flexors/extensors) and at the ankle (dorsiflexors/plantarflexors) were not significantly different between groups or joints (figure 4.7). Correlations were run on the amputee group alone, the control group alone and all participants pooled together. Significance and trends were present in the correlations between the balance variables and cutaneous sensation, proprioception and strength. An example of a correlation is given in figure 4.8. The correlation between RMSv before the perturbation and knee extension strength is significant for the amputee group alone but not for

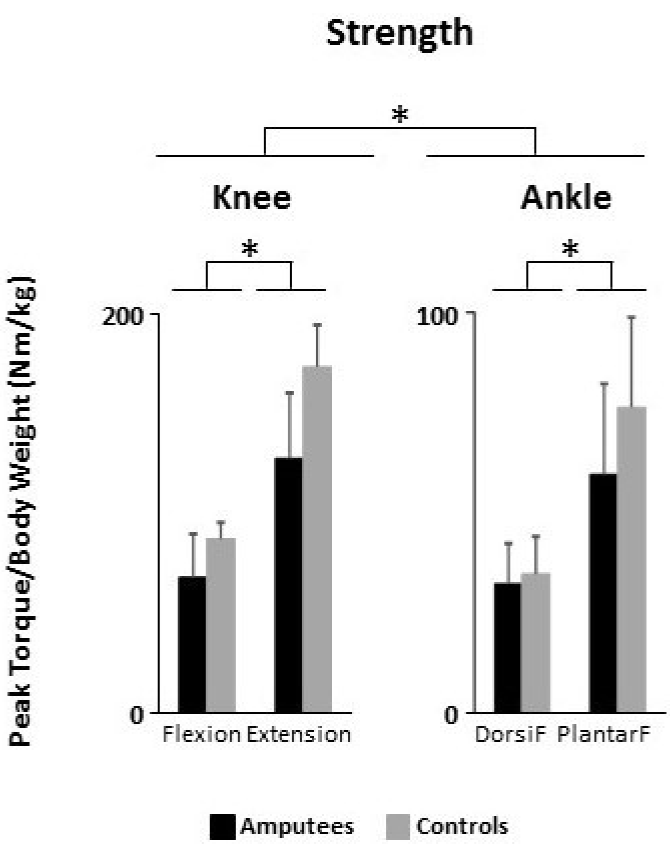

*Figure 4.6. Mean torque normalized to body weight in flexion and extension at the knee, and dorsiflexion and plantarflexion at the ankle in each group. * p < 0.05*

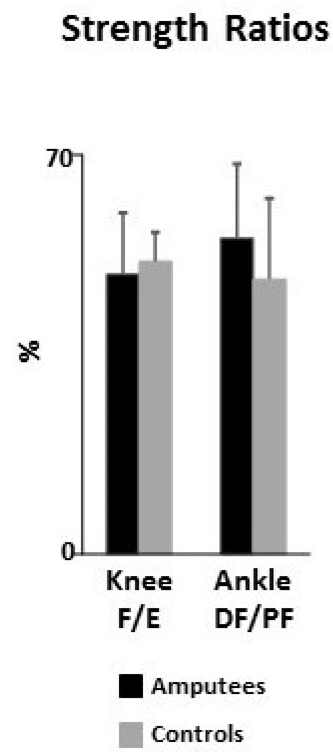

*Figure 4.7. Mean flexors/extensors ratio at the knee and dorsiflexiors/plantarflexors ratio at the ankle in each group. *p<0.05*

all participants pooled together nor for the controls alone. In the amputee group, balance variables were mainly correlated with the strength variables. All significant correlations and trends in the amputee group are presented in Tables 4.1, 4.2 and 4.3. If the correlation was significant in all participants or in the control group only but not in amputees alone, the results are not shown in the tables below. Additional tables including all correlation data are in the appendix.

4.2.3 Discussion

Another aspect of this thesis was to assess cutaneous sensation, proprioception and strength and to compare between groups and to see if a relationship exists between these variables and balance. TPS values were not significantly different between groups or between the tibial site and the plantar aspect. Our results do not agree with the literature both when comparing the groups as well as testing sites. Better sensation has been recorded in the tibial site compared to the plantar aspect in amputees (Kavounoudias et al., 2005). Although Kavounoudias et al. (2005) also reported no difference between groups for the tibial site, our results do not agree regarding sensation at the

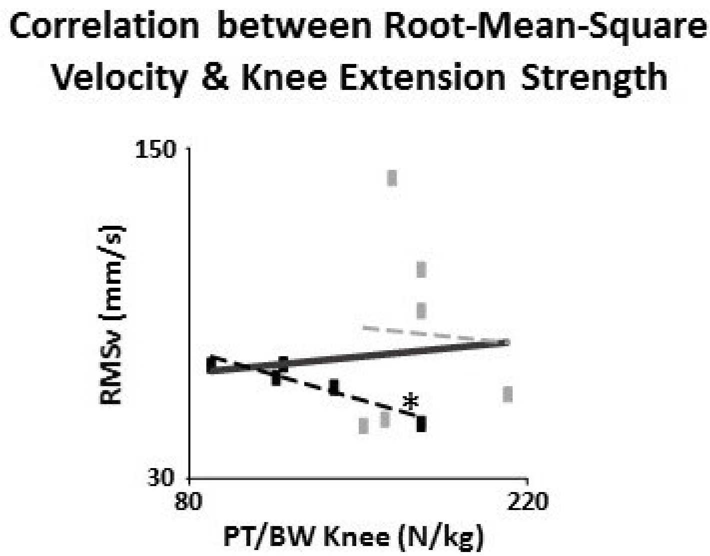

*Figure 4.8. Correlation between COP velocity root-mean-square (RMSv) before the perturbation and knee extension peak torque to body weight (PT/BW). * p<0.05*

plantar aspect when comparing amputees to able-bodied controls. Surprisingly, they observed traumatic amputees to have less sensation in the plantar aspect compared to controls, whereas no differences was reported between vascular amputees and controls. Kavounoudias et al. (2005) hypothesized that adaptations occur to even out the sensory input in the amputated and non-amputated legs.

In our study, there were no differences between groups in the TDPM test, although a trend was shown that participants first felt movement at larger angles in flexion than in extension. The literature is conflicting were Liao and Skinner (1995) found no difference in TDPM between the non-amputated leg of mostly traumatic amputees and controls legs, whereas Kavounoudias et al. (2005) found that regardless of amputee type, amputees have lower proprioception than controls. Our protocol tested TDPM under weight-bearing conditions whereas both Liao and Skinner (1995) and Kavounoudias et al. (2005) tested TDPM without weight bearing which may explain the difference in results.

Many studies have tested strength in the amputated leg, but to our knowledge only one study has looked at strength in the non-amputated leg. Our results regarding strength in amputees are in agreement with the literature with amputees' non-amputated leg being weaker than control legs (Pedrinelli et al., 2002). Also, in our study, both joints and directions tested were significantly different, which lines up with previous literature (Bogdanis and Kalapotharakos, 2016).

As for the relationship between balance and TPS, TDPM and strength, out of the three variables, we observed strength to be the most correlated with balance in amputees. According to our results, an increase in strength at the knee is related to less sway whereas an increase in strength at the ankle in dorsiflexion is related to an increase in sway. Maybe strength at the ankle allows for more control of balance resulting in a more floppy configuration. To our knowledge, the literature only reports on the relationship between balance and strength at the hip. In fact, Nadollek et al. (2002) found that decreased strength in the hip of the amputated leg led to more sway but that the correlation did not hold on the non-amputated leg. We assessed different joints than Nadollek et al. (2002) which may be a reason for our findings not lining up. Nadollek et al. (2002) recruited vascular amputees only, while we studied balance in traumatic amputees. Etiology of amputation has been found to affect sway, which may explain why our results are conflicting

Table 4.1. TPS vs Balance: Significant correlations and trends in amputees between TPS variables and COP and COM. Data for controls and all participants pooled is shown regardless of significance or not (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	TPS	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Tibia	0.859	0.029*	0.218	0.678	0.279	0.380
COM Onset	Tibia	-0.802	0.055†	0.039	0.941	-0.482	0.112
COM Mean Position After	Foot	-0.763	0.078†	0.702	0.120	-0.644	0.024*
COP-COM Onset	Foot	-0.769	0.074†	-0.247	0.637	-0.373	0.232
COP-COM Maximum Amplitude	Tibia	-0.861	0.061†	-0.534	0.275	-0.452	0.162

Table 4.2. TDPM vs Balance: correlation trend in amputees between TDPM variables and COP and COM. Data for controls and all participants pooled is shown regardless of significance or not († trend)

		Amputees		Controls		All Participants	
Balance Variables	TDPM	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
COP-COM Time to Maximum	Extension	-0.928	0.072†	0.419	0.409	-0.427	0.218

(Hermodsson et al., 1994). Time since amputation was also longer in our participants than in the participants in Nadollek et al. (2002), which has been found to affect sway (Mayer et al., 2011).

Table 4.3. Strength vs Balance: Significant correlations and trends in amputees between strength variables and COP and COM. Data for controls and all participants pooled is shown regardless of significance or not (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSv Before	Knee Flexion	-0.918	0.010*	0.137	0.795	0.091	0.780
RMSv Before	Knee Extension	-0.929	0.023*	-0.053	0.920	0.106	0.756
RMSv Before	Ankle Dorsiflexion	0.833	0.080†	0.817	0.091†	0.433	0.211
COM Mean Position Before	Knee Flexion	0.869	0.025*	-0.308	0.553	0.626	0.029*
COM Mean Position Before	Knee Extension	0.911	0.031*	-0.489	0.325	0.524	0.098†
COM Mean Position Before	Ankle Dorsiflexion	0.838	0.076†	0.448	0.450	0.663	0.037*
COM Mean Position After	Knee Flexion	0.748	0.087†	-0.462	0.358	0.451	0.141
COM Mean Position After	Knee Extension	0.870	0.055†	-0.785	0.065†	0.336	0.312
COM Mean Position After	Ankle Dorsiflexion	0.822	0.087†	0.607	0.278	0.704	0.023*
COP Time to Minimum	Knee Extension	-0.967	0.007*	0.541	0.267	-0.802	0.001*
COP Minimum Value	Ankle Ratio DF/PF	0.818	0.091†	0.292	0.634	0.566	0.088†
COP-COM Time to Minimum	Knee Ratio F/E	-0.935	0.020*	0.642	0.169	0.313	0.349
COP-COM Time to Minimum	Ankle Dorsiflexion	-0.915	0.029*	0.414	0.488	-0.237	0.511

4.3 Electromyography

4.3.1 Methods

4.3.1.1 Data Collection & Procedures

Muscle activity was recorded during the balance testing. When preparing for the balance testing, after measuring height, weight and anthropometric measures of the participants, the skin was prepared for electrode placement and the electrodes were affixed. Skin preparation consisted of marking off the locations with a pen after noting the required anatomical landmarks, the area was then shaved, scrubbed with an alcohol swab and dried. Electrodes were affixed to the skin and connected to the receivers. Muscle testing was then performed to ensure a good signal.

EMG was recorded in eight muscles with a sampling rate set at 1500 Hz; rectus abdominis (RA), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA), peroneus longus (PL), erector spinae (ES), biceps femoris (BF), gastrocnemius medialis (GM), using the TeleMyo Direct Transmission System (Noraxon, Scottsdale, AZ, USA). The electrodes were placed according to the SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) placement guidelines. For the RA, the electrode was placed 3cm lateral to the umbilicus. The electrode of the RF was placed halfway between the anterior superior iliac spine, and the top of the patella. For the VM, 80% of the distance between the anterior-superior iliac spine and the medial border of the joint space of the knee was marked off. The electrode for the TA was placed 1/3 of the distance between the line connecting the head of the fibula and the medial malleolus. The PL electrode was placed one quarter of the way between the head of the fibula and the lateral malleolus. The ES electrode was placed two fingers width lateral to the spinous process of L1. The electrode of the BF was placed halfway between the ischial tuberosity and the lateral epicondyle of the tibia. Lastly, the GM was placed on the most prominent part of the muscle following the direction of the muscle fibers.

4.3.1.2 Data Analysis

The EMG signal was put through a 30-350Hz bandpass filter. It was then rectified and a 20 Hz low-pass filter was used to generate the EMG envelopes for each muscle. The onset of muscle contraction was calculated by taking the point at which the envelope surpassed the baseline calculated over 200ms before the onset of the platform translation as the mean plus two standard deviations of the rectified signal. Latency was calculated in relation to the onset of the platform. RMS of the filtered EMG signal was also calculated 8s before (-8 to onset) and 8s after (2 to 8s post onset) the perturbation and was normalized to the maximum amplitude of the signal.

4.3.1.3 Statistical Analysis

The data was averaged across trials for each participant. T-tests were run between groups for the EMG latency data. EMG RMS data was analyzed with 2X2 ANOVAs with a within (pre/post) and a between factor (group). $p<0.05$ was considered significant.

4.3.2 Results

Muscle activity in all eight muscles is represented in figure 4.9. All muscles show clear bursts after the perturbation. Tonic activation before the perturbation, followed by inhibition immediately after the perturbation, occurs in the PL and GM signals. The latencies in all muscles are not different between groups except for a trend in PL with the amputees contracting earlier (Figure 4.10; $t_{5.810}=2.015$, $p=0.092$). The latencies in GM and PL occur later than in the other muscles due to inhibition that occurred before the burst. Inhibition of the GM appeared in all participants. PL on the other hand had two distinctive patterns, with 5/6 controls exhibiting inhibition and 5/6 amputees displaying a burst pattern (Figure 4.11).

Increases in RMS values after the perturbation were seen in 5 muscles (Figure 4.12; RA: $F_{1,10}=30.752$, $p<0.001$; RF: $F_{1,10}=16.370$, $p=0.002$; VM: $F_{1,10}=6.688$, $p=0.027$; TA: $F_{1,10}=27.537$, $p<0.001$; ES: $F_{1,10}=9.698$, $p=0.011$). There were no significant differences between groups in any muscle, nor were there group*time interactions.

4.3.3 Discussion

In our study, RMS values increased after the perturbation. Co-contraction may have contributed to the increase in RMS values in order to maintain balance in a more difficult situation. It has been shown that amputees co-contract during gait (Seyedali et al., 2012). However, there were no differences in RMS values between groups both before and after the perturbation in our study. It is also possible that the increase in RMS indicates that the perturbation may have induced muscular fatigue (Cao et al., 2017; Oberg, 1996). We also observed no difference in onset of muscle activation between groups. This is in agreement with Viton et al. (2000) who reported that muscle activity onset in the GM is no different in amputees compared to controls during lateral leg raising. However, Rusaw et al. (2013) observed later muscle activation onsets in amputees compared to controls during support surface tilting in the GM of the non-amputated leg and the BF of the amputated leg. We also reported a different pattern in the PL, where most amputees would contract the PL in response to the perturbation whereas most controls would inhibit the PL first before recruiting it later. This agrees with the literature where activation patterns are also different in amputees compared to controls with amputees using the tensor fascia latae to initiate COP shift before lateral leg raising whereas controls recruit the GM (Viton et al., 2000)

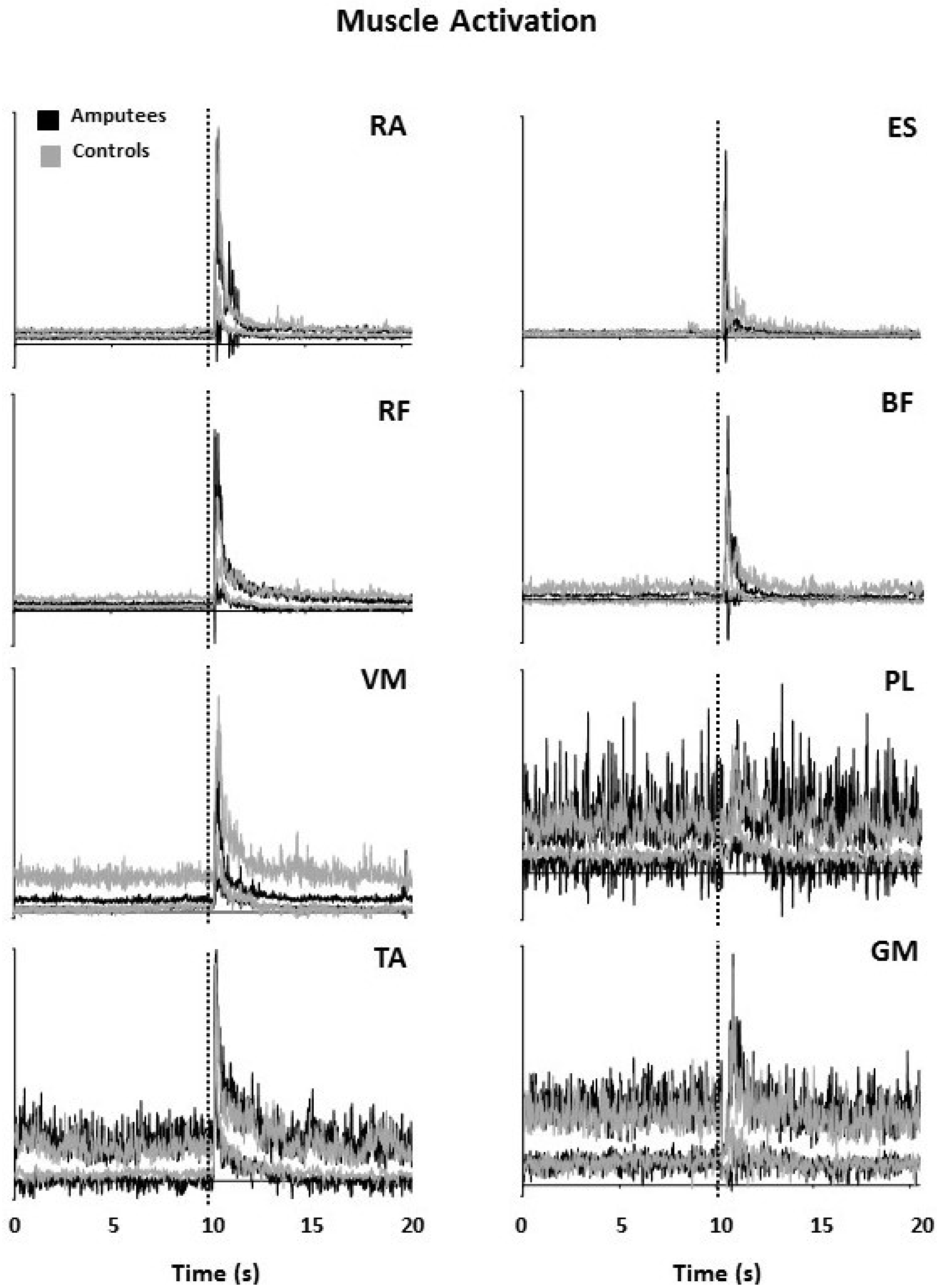

Figure 4.9. Mean muscle activation patterns for each group, for rectus abdominis (RA), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA), peroneus longus (PL), erector spinae (ES), biceps femoris (BF), and gastrocnemius medialis (GM) for each group. The lines display one standard deviation above and below the mean. All trials were aligned with the perturbation onset fixed at 10s. The vertical dotted line represents the onset of the perturbation.

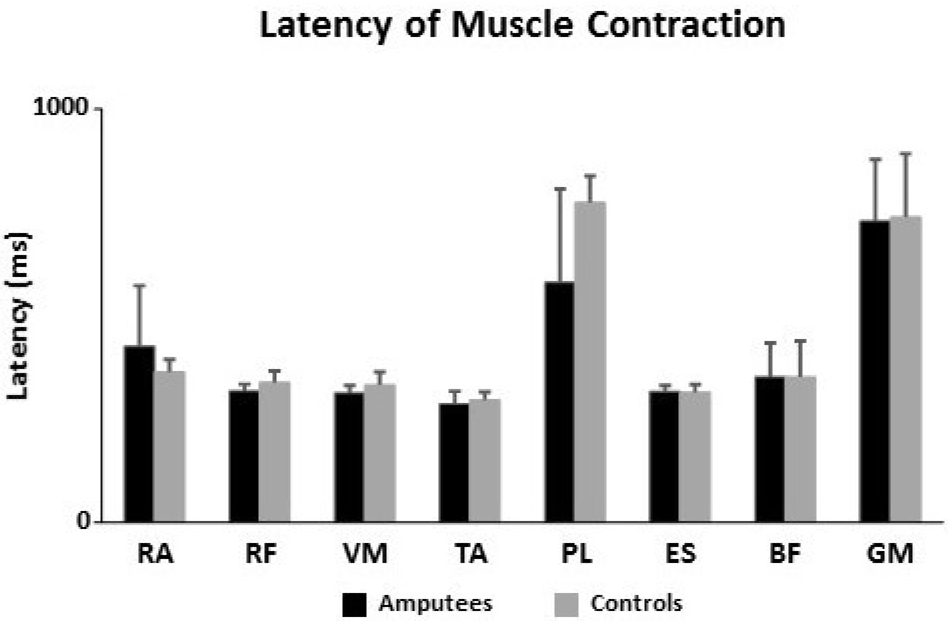

Figure 4.10. Mean latencies of muscle contraction for RA, RF, VM, TA, PL, ES, BF, and GM in each group.

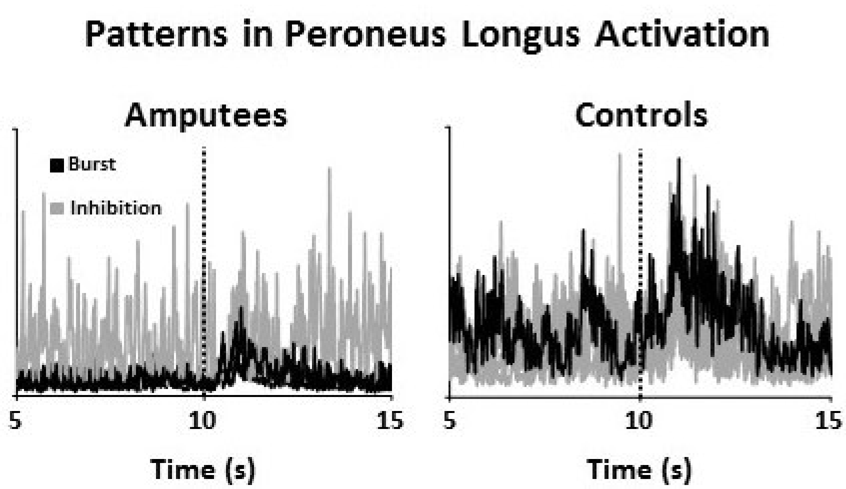

Figure 4.11. Different patterns in peroneus longus activation in each group

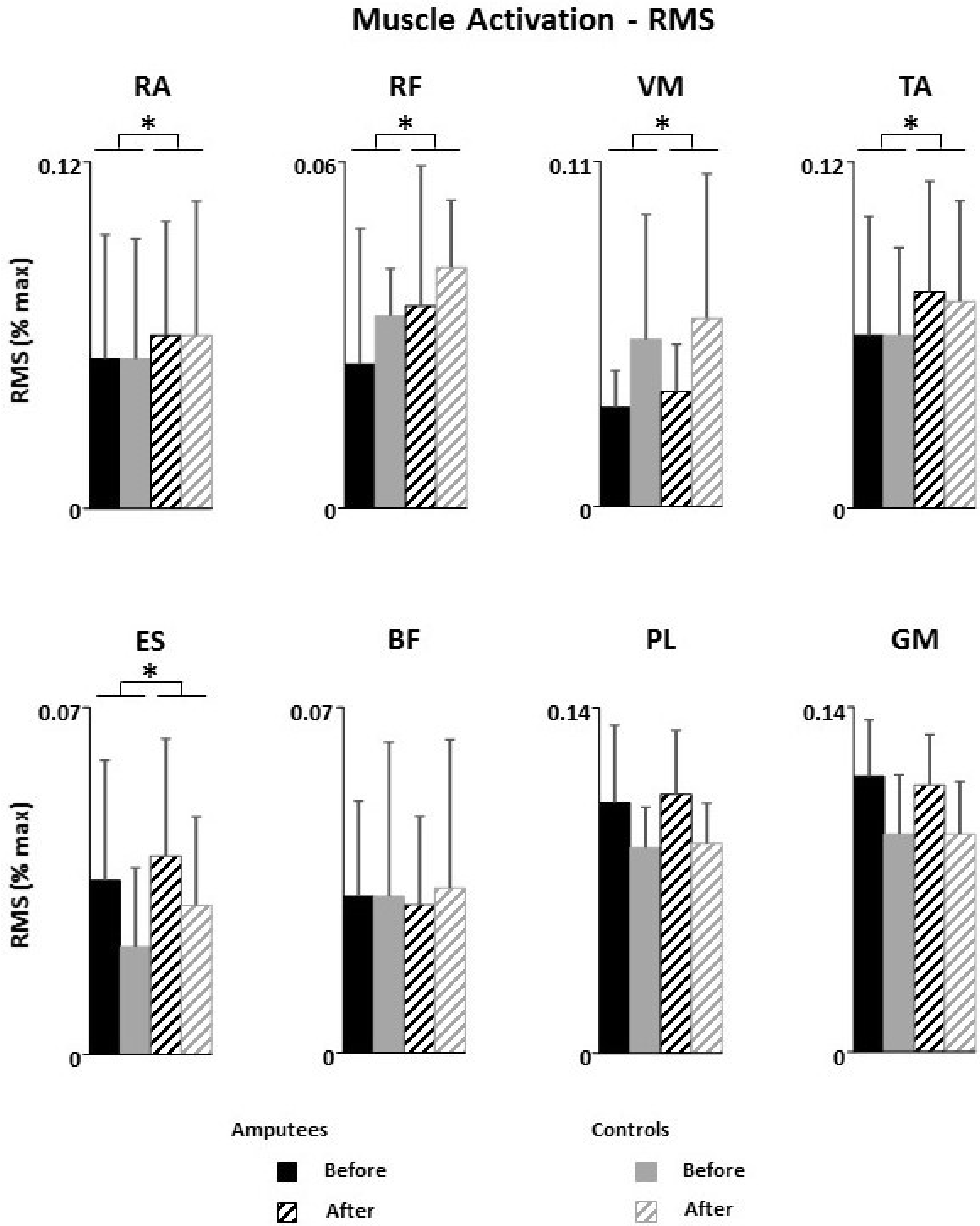

Figure 4.12. Mean RMS of the electromyographic signal before and after the perturbation in each group.
** p<0.05*

CHAPTER 5 – CONCLUDING REMARKS

Goal 1. The first goal was to assess balance in amputees during single-legged stance on the non-amputated leg during perturbed conditions by evaluating center of pressure (COP) and center of mass (COM). Electromyography (EMG) was also measured to assess muscular contribution to balance.

Hypothesis 1.1. We hypothesized that the COP movement before the perturbation would be the same in both groups and would increase after the perturbation, with a larger increase in amputees. There was no difference in sway between both groups which agrees with the literature stating no difference during quiet standing (Curtze et al., 2016; Hermodsson et al., 1994). The only variable with a trend in group*time interaction was the average center of pressure velocity (COPv), indicating increased sway in controls after the perturbation. Although contrary to our hypothesis, this could be explained by a more floppy reaction in controls since they are not faced with the same fear of falling present in amputees (Miller et al., 2001a).

Hypothesis 1.2. During the perturbation, we expected no difference in COP and COM peak displacement between groups. The COP and COM peak displacements were not different between groups which is in agreement with the literature (Bolger et al., 2014). The latency of the peak was faster in amputees which is in agreement with the literature (Molina-Rueda et al., 2016).

Hypothesis 1.3. We hypothesized that EMG activation patterns would be different between groups. Onsets of muscle activation were not different between groups although the pattern within the PL differed between groups with amputees adopting a "burst" reaction to the perturbation and controls adopting an "inhibition" reaction. Our results are in disagreement with Rusaw et al. (2013) who found increased latency in the gastrocnemius medialis (GM) and biceps femoris (BF) of amputees during support surface tipping. However, similar to our study, different activation patterns have been shown in amputees and controls during lateral leg raising with amputees recruiting the tensor fascia latae and controls recruiting the GM (Viton et al., 2000).

Hypothesis 1.4 We expected an increase in EMG root-mean-square (RMS) values after the perturbation in both groups with a larger increase in amputees. Increased RMS after the perturbation was found in 5 muscles; rectus abdominis (RA), rectus femoris (RF), vastus medialis (VM), tibialis anterior (TA) and erector spinae (ES). However, no difference in RMS values was found between groups. The increase in RMS after the perturbation may be due to fatigue or co-contractions (Cao et al., 2017; Oberg, 1996).

Goal 2. The second goal was to assess cutaneous sensation, proprioception and strength in the non-amputated leg.

Hypothesis 2.1. We hypothesized that cutaneous sensation would be decreased in amputees compared to controls. In our study, cutaneous sensation was found not to be different in both groups both at the tibial site and the plantar aspect. This goes against the literature stating that amputees have decreased sensation compared to controls at the plantar aspect, although not at the tibial site (Kavounoudias et al., 2005).

Hypothesis 2.2. We hypothesized that proprioception would be decreased in amputees compared to controls. The results of our study indicate no difference between groups. The literature is conflicting, where Kavounoudias et al. (2005) observed decreased proprioception in the non-amputated leg and Liao and Skinner (1995) found no difference between the non-amputated leg and control legs.

Hypothesis 2.3. Our hypothesis for the strength test was that the non-amputated leg would be weaker than control legs. Our results revealed a trend for amputees to be weaker than controls at the knee and ankle which agrees with previous work at the knee (Pedrinelli et al., 2002). In Isakov et al. (1996a), there is a trend for increased time since amputation leading to decreased strength due to atrophy. This may be why we only have a trend with our amputees having been amputated nine year prior to testing.

Goal 3. Our last goal was to study the role of cutaneous sensation proprioception, and strength and how they relate to balance as measured by COP and COM variables.

Hypothesis 3.1. We hypothesized that there would be a relationship between balance and cutaneous sensation, proprioception and strength and that a decrease in these three variables would lead to a decrease in balance abilities. To our knowledge, correlations in amputees have only been run on cutaneous sensation and strength (Nadollek et al., 2002; Quai et al., 2005). Based on our results, strength was the element most often significantly correlated to balance variables. We found that an increase in strength at the knee led to a decrease in sway and that an increase in strength at the ankle led to an increase in sway. Our results are difficult to compare to the literature where an increase in strength at the hip led to a decrease in sway (Nadollek et al., 2002). An increase in strength at the ankle may mean more balance confidence, which may lead to an increase in sway.

CHAPTER 6 − REFERENCES

Anker, L.C., Weerdesteyn, V., van Nes, I.J., Nienhuis, B., Straatman, H., Geurts, A.C., 2008. The relation between postural stability and weight distribution in healthy subjects. Gait Posture 27, 471-477.
Blaszczyk, J.W., Beck, M., Sadowska, D., 2014. Assessment of postural stability in young healthy subjects based on directional features of posturographic data: vision and gender effects. Acta Neurobiol Exp 74, 433-442.
Blaszczyk, J.W., Orawiec, R., Duda-Klodowska, D., Opala, G., 2007. Assessment of postural instablity in patients with parkinsons disease. Exp Brain Res 183, 107-114.
Bogdanis, G.C., Kalapotharakos, V.I., 2016. Knee Extension Strength and Hamstrings-to-Quadriceps Imbalances in Elite Soccer Players. Int J Sports Med 37, 119-124.
Bolger, D., Ting, L.H., Sawers, A., 2014. Individuals with transtibial limb loss use interlimb force asymmetries to maintain multi-directional reactive balance control. Clin Biomech.
Briggs, K.K., Lysholm, J., Tegner, Y., G., R.W., Kocher, M.S., Steadman, R., 2009. The Reliability, Validity, and Reponsiveness of the Lysholm Score and Tegner Activity Scale for Anterior Cruciate Ligament Injuries of the Knee. Am J Sports Med 37, 890-897.
Buckley, J.G., O'Driscoll, D., Bennett, S.J., 2002. Postural sway and active balance performance in highly active lower-limb amputees. Am J Phys Med Rehabil 81, 13-20.
Cao, L., Wang, Y., Hao, D., Rong, Y., Yang, L., Zhang, S., Zheng, D., 2017. Effects of Force Load, Muscle Fatigue, and Magnetic Stimulation on Surface Electromyography during Side Arm Lateral Raise Task: A Preliminary Study with Healthy Subjects. Biomed Res Int 2017, 8943850.
Collins, S., Visscher, P., De Vet, H.C., Zuurmond, W.W., Perez, R.S., 2010. Reliability of the Semmes Weinstein Monofilaments to measure coetaneous sensibility in the feet of healthy subjects. Disabil Rehabil 32, 2019-2027.
Côté, J.N., Patenaude, I., St-Onge, N., Fung, J., 2009. Whiplash associated disorder affect postural reactions to antero-posterior supports surface translations during sitting Gait Posture 29, 603-611.
Curtze, C., Hof, A.L., Postema, K., Otten, B., 2012. The relative contributions of the prosthetic and sound limb to balance control in unilateral transtibial amputees. Gait Posture 36, 276-281.
Curtze, C., Hof, A.L., Postema, K., Otten, B., 2016. Staying in dynamic balance on a prosthetic limb: A leg to stand on? Med Eng Phys 38, 576-580.
Danis, C.G., Krebs, D.E., Gill-Body, K.M., Sahrmann, S., 1998. Relationship Between Standing Posture and Stability. Physical Therapy 78, 502-517.
Davidson, M., de Morton, N., 2006. A Systematic Review of the Human Activity Profile. Clin Rehabil 21, 151-162.
Deshpande, N., Connelly, D.M., Culham, E.G., Costigan, P.A., 2003. Reliability and Validity of Ankle Proprioceptive Measures. Arch Phys Med Rehabil 84, 883-889.
Dillingham, T.R., Pezzin, L.E., MacKenzie, E.J., 1998. Incidence, acute care length of stay, and discharge to rehabilitation of traumatic amputee patients: an epidemiologic study. Arch Phys Med Rehabil 79, 279-287.

Eakin, C.L., Quesada, P.M., Skinner, H., 1992. Lower-limb proprioception in above-knee amputees. Clin Orthop Relat Res (284), 239-246.
Ebskov, L.B., 1992. Level of lower limb amputation in relation to etiology: an epidemiological study. Prosthet Orthot Int 16, 163-167.
Fraisse, N., Martinet, N., Kpadonou, T.J., Paysant, J., Blum, A., André, J.M., 2008. Les muscles de l'amputé tibial. Ann Readapt Med Phys 51, 218-227.
Gaines, J.M., Talbot, L.A., 1999. Isokinetic strength testing in research and in practice. Biol Res Nurs 1, 57-64.
Gauthier-Gagnon, C., St-Pierre, D., Drouin, G., Riley, E., 1986. Augmented sensory feedback in the early training of standing balance of below-knee amputees. Physiother Can 38, 137-142.
Gregory-Dean, A., 1991. Amputations: statistics and trends. Ann R Coll Surg Engl 73, 137-142.
Hall, J.E., Guyton, A.C., 2011. Guyton and Hall textbook of medical physiology, 12 ed. Saunders Elsevier, Philadelphia, PA.
Harden, R.N., Gagnon, C.M., Khan, A., Wallach, G., Zereshki, A., 2010. Hypoesthesia in the Distal Residual Limb of Amputees. PM&R 2, 607-611.
Hermodsson, Y., Ekdahl, C., Persson, B.M., Roxendal, G., 1994. Standing balance in trans-tibial amputees following vascular disease or trauma: a comparative study with healthy subjects. Prosthet Orthot Int 18, 150-158.
Hlavackova, P., Franco, C., Diot, B., Vuillerme, N., 2011. Contribution of each leg to the control of unperturbed bipedal stance in lower limb amputees: new insights using entropy. PloS one 6, e19661.
Horak, F.B., Nutt, J.G., Nashner, L.M., 1992. Postural inflexibility in parkinsonian subjects. J Neurol Sci 111, 46-58.
Isakov, E., Burger, H., Gregoric, M., Marincek, C., 1996a. Isokinetic and isometric strength of the thigh muscles in below-knee amputees. Clin Biomech 11, 233-235.
Isakov, E., Burger, H., Gregoric, M., Marincek, C., 1996b. Stump length as related to atrophy and strength of the thigh muscles in trans-tibial amputees. Prosthet Orthot Int 20, 96-100.
Isakov, E., Mizrahi, J., Ring, H., Susak, Z., Hakim, N., 1992. Standing sway and weight-bearing distribution in people with below-knee amputations. Arch Phys Med Rehabil 73, 174-178.
Jayakaran, P., Johnson, G.M., Sullivan, S.J., Nitz, J.C., 2012. Instrumented measurement of balance and postural control in individuals with lower limb amputation: a critical review. Int J Rehabil Res 35, 187-196.
Jerosch-Herold, C., 2005. Assessment of Sensibility after Nerve Injury and Repair: A Systematic Review of Evidence for Validity, Reliability and Responsiveness of Tests. J Hand Surg 30B, 252-264.
Kanade, R.V., Van Deursen, R.W., Harding, K.G., Price, P.E., 2008. Investigation of standing balance in patients with diabetic neuropathy at different stages of foot complications. Clin Biomech (Bristol, Avon) 23, 1183-1191.
Kavounoudias, A., Tremblay, C., Gravel, D., Iancu, A., Forget, R., 2005. Bilateral changes in somatosensory sensibility after unilateral below-knee amputation. Arch Phys Med Rehabil 86, 633-640.

Kayssi, A., de Mestral, C., Forbes, T.L., Roche-Nagle, G., 2016a. A Canadian population-based description of the indications for lower-extremity amputations and outcomes. Can J Surg 59, 99-106.
Kayssi, A., Dilkas, S., Dance, D.L., de Mestral, C., Forbes, T.L., Roche-Nagle, G., 2016b. Rehabilitation Trends After Lower Extremity Amputations in Canada. PM R.
Kosasih, J.B., Silver-Thorn, M., 1998. Sensory Changes in Adults with Unilateral Transtibial. J Rehabil Res Dev 35, 85.
Ku, P.X., Abu Osman, N.A., Wan Abas, W.A.B., 2014. Balance control in lower extremity amputees during quiet standing: A systematic review. Gait Posture 39, 672-682.
Kulkarni, J., Wright, S., Toole, C., Morris, J., Hirons, R., 1996. Falls in Patients with Lower Limb Amputations: Prevalence and Contributing Factors. Physiother 82, 130-136.
Latanioti, E.P., Angoules, A.G., Boutsikari, E.C., 2013. Proprioception in above-the-knee amputees with artificial limbs. Scientific World J 2013, 417982.
Lenka, P., Tiberwala, D.N., 2010. Effect of stump length on postural steadiness during quiet stance in unilateral trans-tibial amputees. Al Ameen J Med Sci 1, 50-57.
Liao, K.I., Skinner, H.B., 1995. Knee joint proprioception in below-knee amputees. Am J Knee Surg 8, 105-109.
Mandel, A., MscOt, Paul, K., MscOt, Paner, R., Devlin, M., Dilkas, S., Pauley, T., 2016. Balance confidence and activity of community-dwelling patients with transtibial amputation. J Rehabil Res Dev 53, 551-560.
Mayer, A., Tihanyi, J., Bretz, K., Csende, Z., Bretz, E., Horvath, M., 2011. Adaptation to altered balance conditions in unilateral amputees due to atherosclerosis: a randomized controlled study. BMC Musculoskelet Disord 12, 118.
Miller, W.C., Deathe, A.B., 2011. The influence of balance confidence on social activity after discharge from prosthetic rehabilitation for first lower limb amputation. Prosthet Orthot Int 35, 379-385.
Miller, W.C., Deathe, A.B., Speechley, M., Koval, J., 2001a. The influence of falling, fear of falling, and balance confidence on prosthetic mobility and social activity among individuals with a lower extremity amputation. Arch Phys Med Rehabil 82, 1238-1244.
Miller, W.C., Speechley, M., Deathe, B., 2001b. The prevalence and risk factors of falling and fear of falling among lower extremity amputees. Arch Phys Med Rehabil 82, 1031-1037.
Mohieldin, A.H.A., Chidambaram, A., Sabapathivinayagam, R., Al Busairi, W., 2010. Quantitative Assessment of Postural Stability and Balance Between Persons with Lower Limb Amputation and Normal Subjects by using Dynamic Posturography. Maced J Med Sci 3, 138-143.
Moirenfeld, I., Ayalon, M., Ben-Sira, D., Isakov, E., 2000. Isokinetic strength and endurance of the knee extensors and flexors in trans-tibial amputees. Prosthet Orthot Int 24, 221-225.
Molina-Rueda, F., Molero-Sanchez, A., Alguacil-Diego, I.M., Carratala-Tejada, M., Cuesta-Gomez, A., Miangolarra-Page, J.C., 2016. Weight Symmetry and Latency

Scores for Unexpected Surface Perturbations in Subjects With Traumatic and Vascular Unilateral Transtibial Amputation. PM R 8, 235-240.
Mouchnino, L., Mille, M.L., Cincera, M., Bardot, A., Delarque, A., Pedotti, A., Massion, J., 1998. Postural reorganization of weigh-shifting in below-knee amputees during leg raising. Exp Brain Res 121, 205-214.
Moxey, P.W., Hofman, D., Hinchliffe, R.J., Jones, K., Thompson, M.M., Holt, P.J., 2010. Epidemiological study of lower limb amputation in England between 2003 and 2008. Br J Surg 97, 1348-1353.
Nadollek, H., Brauer, S., Isles, R., 2002. Outcomes after trans-tibial amputation: the relationship between quiet stance ability, strength of hip abductor muscles and gait. Physiother Res Int 7, 203-214.
Nagai, T., Sell, T.C., Abt, J.P., Lephart, S.M., 2012. Reliability, precision, and gender differences in knee internal/external rotation proprioception measurements. Phys Ther Sport 13, 233-237.
Nederhand, M.J., Van Asseldonk, E.H., van der Kooij, H., Rietman, H.S., 2012. Dynamic Balance Control (DBC) in lower leg amputee subjects; contribution of the regulatory activity of the prosthesis side. Clin Biomech (Bristol, Avon) 27, 40-45.
Nolan, L., 2009. Lower limb strength in sports-active transtibial amputees. Prosthet Orthot Int 33, 230-241.
Oberg, T., 1996. Muscle fatigue and calibration of EMG measurements. J. Electromyogr. Kinesiol. 5, 239-243.
Opila, K.A., Wagner, S.S., Schiowitz, S., Chen, J., 1988. Postural alignment in barefoot and high-heeled stance. Spine 13, 542-547.
Ownings, M., Kozak, L., 1998. Ambulatory and inpatient procedures in the United States, 1996. National Health Statistics. Vital Health Stat 13.
Pedrinelli, A., Saito, M., Coelho, R.F., Fontes, R.B., Guarniero, R., 2002. Comparative study of the strength of the flexor and extensor muscles of the knee through isokinetic evaluation in normal subjects and patients subjected to trans-tibial amputation. Prosthet Orthot Int 26, 195-205.
Pezzin, L.E., Dillingham, T.R., MacKenzie, E.J., 2000. Rehabilitation and the long-term outcomes of persons with trauma-related amputations. Arch Phys Med Rehabil 81, 292-300.
Purves, D., Augustine, G.J., Fitzpatrick, D., Hall, W.C., LaMantia, A., McNamara, J.O., Williams, S.M., 2004. NEUROSCIENCE, Third Edition ed. Sinauer Associates, Sunderland, MA.
Quai, T.M., Brauer, S.G., Nitz, J.C., 2005. Somatosensation, circulation and stance balance in elderly dysvascular transtibial amputees. Clin Rehabil 19, 668-676.
Renstrom, P., Grimby, G., Larsson, E., 1983a. Thigh muscle strength in below-knee amputees. Scand J Rehabil Med 9, 163-173.
Renstrom, P., Grimby, G., Morelli, B., Palmertz, B., 1983b. Thigh muscle atrophy in below-knee amputees. Scand J Rehabil Med.
Rougier, P.R., Bergeau, J., 2009. Biomechanical Analysis of Postural Control of Persons with Transtibial or Transfemoral Amputation. Am J Phys Med Rehabil 88, 896-903.

Rusaw, D., Hagberg, K., Nolan, L., Ramstrand, N., 2013. Bilateral electromyogram response latency following platform perturbation in unilateral transtibial prosthesis users: Influence of weight distribution and limb position. J Rehabil Res Dev 50, 531.
SENIAM website, Recommendations for sensor locations on individual muscles.
Seyedali, M., Czerniecki, J.M., Morgenroth, D.C., Hahn, M.E., 2012. Co-contraction patterns of trans-tibial amputee ankle and knee musculature during gait. J Neuroeng Rehabil 9.
Tegner, Y., Lysholm, J., 1985. Rating Systems in the Evaluation of Knee Ligament Injuries. Clin Orthop Relat Res &NA;, 43-49.
Termoz, N., Martin, L., Prince, F., 2004. Assessment of postural response after a self initiated perturbation. Motor Control 8, 51-63.
Tugcu, I., Safaz, I., Yilmaz, B., Goktepe, A.S., Taskaynatan, M.A., Yazicioglu, K., 2009. Muscle strength and bone mineral density in mine victims with transtibial amputation. Prosthet Orthot Int 33, 299-306.
Vicon Motion Systems, 2010. Plug In Gait Product Guide -- Foundation Notes.
Viton, J.M., Mouchnino, L., Mille, M.L., Cincera, M., Delarque, A., Pedotti, A., Bardot, A., Massion, J., 2000. Equilibrium and movement control strategies in trans-tibial amputees. Prosthet Orthot Int 24, 108-116.
Vittas, D., Larsen, T.K., Jansen, E.C., 1986. Body sway in below-knee amputees. Prosthet Orthot Int 10, 139-141.
Vrieling, A.H., van Keeken, H.G., Schoppen, T., Otten, E., Hof, A.L., Halbertsma, J.P.K., Postema, K., 2008. Balance control on a moving platform in unilateral lower limb amputees. Gait Posture 28, 222-228.
Winter, D.A., 1995. Human balance and posture control during standing and walking. Gait Posture 3, 193-214.
Winter, D.A., Patla, A.E., Prince, F., Ishac, M., Gielo-Perczak, K., 1998. Stiffness control of balance in quiet standing. J. Neurophysiol 80, 1211-1221.
Ziegler-Graham, K., MacKenzie, E.J., Ephraim, P.L., Travison, T.G., Brookmeyer, R., 2008. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Arch Phys Med Rehabil 89, 422-429.

APPENDIX A: Correlation Tables

Table A.1.a. Correlations between touch pressure sensation (TPS) at the tibia & center of pressure (COP) and center of mass (COM) (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	TPS	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Tibia	.859	.029*	.103	.845	.193	.547
RMSd After	Tibia	.149	.778	.708	.116	.352	.262
COPv Before	Tibia	.043	.935	.324	.531	.103	.749
COPv After	Tibia	.245	.640	.704	.119	.304	.337
RMSv Before	Tibia	.498	.315	.256	.625	.118	.716
RMSv After	Tibia	.344	.504	.712	.113	.360	.251
COM Mean Position Before	Tibia	-.542	.267	-.462	.356	-.529	.077†
COM Mean Position After	Tibia	-.413	.416	-.469	.349	-.420	.174
COP Min Amplitude	Tibia	-.110	.836	-.057	.915	.007	.982
COM Min Amplitude	Tibia	-.111	.834	-.451	.369	-.272	.393
COP-COM Min Amplitude	Tibia	-.385	.451	-.118	.824	-.264	.408
COP-COM Max Amplitude	Tibia	-.861	.061†	-.530	.279	-.490	.126
COP Min Latency	Tibia	.514	.296	.449	.372	.517	.085†
COM Min Latency	Tibia	.040	.940	.458	.361	.229	.475
COP-COM Min Latency	Tibia	.197	.709	-.541	.268	-.323	.306
COP-COM Max Latency	Tibia	.169	.786	.334	.517	.248	.462
COP Onset	Tibia	-.164	.756	.747	.088†	.336	.286
COM Onset	Tibia	-.802	.055†	.149	.778	-.200	.534
COP-COM Onset	Tibia	-.136	.798	.177	.737	.166	.607
COP Offset	Tibia	.179	.734	.497	.316	.314	.321
COM Offset	Tibia	-.327	.527	-.046	.932	-.149	.643
COP-COM Offset	Tibia	.527	.283	.003	.995	.349	.266

Table A.1.b Correlations between TPS at the foot & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	TPS	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Foot	.571	.237	-.201	.703	.007	.983
RMSd After	Foot	-.059	.911	.442	.380	.275	.387
COPv Before	Foot	.302	.561	.077	.884	.110	.733
COPv After	Foot	.188	.721	.648	.164	.434	.159
RMSv Before	Foot	.582	.226	-.050	.924	.034	.915
RMSv After	Foot	.400	.433	.581	.226	.441	.151
COM Mean Position Before	Foot	-.716	.109	-.432	.392	-.525	.079†
COM Mean Position After	Foot	-.763	.078†	-.505	.307	-.608	.036*
COP Min Amplitude	Foot	-.414	.414	-.275	.598	-.320	.311
COM Min Amplitude	Foot	-.658	.155	-.574	.234	-.572	.052†
COP-COM Min Amplitude	Foot	.076	.886	.234	.656	.097	.765
COP-COM Max Amplitude	Foot	-.493	.399	-.245	.640	-.306	.360
COP Min Latency	Foot	.288	.579	.202	.701	.210	.511
COM Min Latency	Foot	-.608	.200	.057	.914	-.313	.322
COP-COM Min Latency	Foot	.317	.540	-.563	.244	-.120	.710
COP-COM Max Latency	Foot	-.332	.585	.280	.592	-.017	.962
COP Onset	Foot	-.593	.214	.657	.156	.223	.486
COM Onset	Foot	-.615	.194	.105	.843	-.248	.438
COP-COM Onset	Foot	-.769	.074†	-.015	.977	-.318	.314
COP Offset	Foot	-.429	.396	.269	.607	-.064	.843
COM Offset	Foot	-.416	.412	-.427	.398	-.422	.171
COP-COM Offset	Foot	-.009	.987	-.254	.628	-.087	.788

Table A.2.a. Correlations between threshold detection of passive motion (TDPM) in flexion & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	TDPM	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Flexion	.726	.165	.037	.945	.186	.585
RMSd After	Flexion	.636	.249	.473	.343	.511	.108
COPv Before	Flexion	-.477	.417	.224	.670	.163	.632
COPv After	Flexion	-.003	.997	.623	.186	.543	.084
RMSv Before	Flexion	.003	.997	.097	.855	.136	.690
RMSv After	Flexion	.647	.238	.564	.243	.569	.068†
COM Mean Position Before	Flexion	-.128	.838	-.187	.722	-.093	.786
COM Mean Position After	Flexion	-.292	.634	-.462	.356	-.320	.338
COP Min Amplitude	Flexion	.486	.406	.022	.967	.132	.699
COM Min Amplitude	Flexion	.092	.883	-.270	.605	-.056	.871
COP-COM Min Amplitude	Flexion	-.597	.287	-.132	.804	-.297	.375
COP-COM Max Amplitude	Flexion	-.027	.973	-.117	.825	-.110	.763
COP Min Latency	Flexion	.216	.727	-.208	.692	-.090	.792
COM Min Latency	Flexion	-.007	.991	.114	.830	-.018	.958
COP-COM Min Latency	Flexion	-.691	.197	-.747	.088†	-.482	.134
COP-COM Max Latency	Flexion	-.833	.167	.438	.385	.062	.865
COP Onset	Flexion	.362	.549	.825	.043*	.713	.014*
COM Onset	Flexion	-.194	.755	-.336	.516	-.267	.428
COP-COM Onset	Flexion	-.363	.548	-.492	.322	-.443	.173
COP Offset	Flexion	-.633	.252	.052	.922	-.090	.792
COM Offset	Flexion	.194	.754	-.358	.486	-.206	.544
COP-COM Offset	Flexion	.634	.251	-.347	.500	-.008	.982

Table A.2.b. Correlations between TDPM in extension & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	TDPM	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Extension	.749	.145	-.210	.689	.078	.820
RMSd After	Extension	.776	.123	.354	.491	.200	.555
COPv Before	Extension	-.461	.434	.051	.923	-.219	.517
COPv After	Extension	.027	.965	.589	.219	.016	.962
RMSv Before	Extension	.040	.949	-.126	.812	-.146	.669
RMSv After	Extension	.725	.165	.469	.348	.158	.642
COM Mean Position Before	Extension	-.138	.825	-.541	.267	-.295	.379
COM Mean Position After	Extension	-.289	.637	-.684	.134	-.382	.247
COP Min Amplitude	Extension	.557	.330	-.340	.509	.288	.391
COM Min Amplitude	Extension	.178	.775	-.682	.136	-.085	.805
COP-COM Min Amplitude	Extension	-.745	.149	.222	.673	-.556	.076†
COP-COM Max Amplitude	Extension	.006	.994	-.053	.921	.111	.759
COP Min Latency	Extension	.337	.580	-.076	.886	.332	.319
COM Min Latency	Extension	.176	.778	.038	.943	.233	.490
COP-COM Min Latency	Extension	-.757	.138	-.695	.126	-.731	.011*
COP-COM Max Latency	Extension	-.928	.072†	.419	.409	-.427	.218
COP Onset	Extension	.499	.392	.611	.198	.295	.378
COM Onset	Extension	-.164	.792	.156	.768	-.030	.931
COP-COM Onset	Extension	-.198	.750	-.077	.885	-.029	.931
COP Offset	Extension	-.472	.423	.202	.700	-.272	.418
COM Offset	Extension	.377	.531	-.413	.415	.059	.863
COP-COM Offset	Extension	.768	.130	-.177	.737	.579	.062†

Table A.3.a. Correlations between strength at the knee in flexion & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength - Knee	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Flexion	-.604	.204	.123	.816	-.052	.872
RMSd After	Flexion	.020	.970	.312	.547	.325	.303
COPv Before	Flexion	-.500	.312	.337	.514	.167	.604
COPv After	Flexion	.050	.926	.679	.138	.470	.123
RMSv Before	Flexion	-.918	.010*	.137	.795	.091	.780
RMSv After	Flexion	-.188	.722	.516	.295	.356	.255
COM Mean Position Before	Flexion	.869	.025*	-.308	.553	.626	.029*
COM Mean Position After	Flexion	.748	.087†	-.462	.357	.451	.141
COP Min Amplitude	Flexion	.375	.464	-.457	.363	-.022	.947
COM Min Amplitude	Flexion	.483	.332	-.385	.451	.336	.285
COP-COM Min Amplitude	Flexion	.221	.674	.073	.890	.253	.428
COP-COM Max Amplitude	Flexion	.517	.372	-.061	.908	.071	.837
COP Min Latency	Flexion	-.720	.107	-.318	.539	-.721	.008*
COM Min Latency	Flexion	.153	.772	-.355	.490	-.126	.697
COP-COM Min Latency	Flexion	-.681	.136	-.310	.549	-.144	.655
COP-COM Max Latency	Flexion	.162	.795	-.041	.939	.071	.836
COP Onset	Flexion	.469	.348	.559	.249	.487	.108
COM Onset	Flexion	.398	.435	-.181	.732	.109	.735
COP-COM Onset	Flexion	.278	.593	-.707	.116	-.198	.537
COP Offset	Flexion	-.267	.609	-.526	.284	-.243	.446
COM Offset	Flexion	.088	.868	-.716	.110	-.094	.772
COP-COM Offset	Flexion	-.223	.671	-.633	.178	-.443	.150

Table A.3.b Correlations between strength at the knee in extension & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength - Knee	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Extension	-.722	.169	.040	.941	-.044	.897
RMSd After	Extension	-.682	.205	.254	.627	.280	.405
COPv Before	Extension	-.427	.474	.093	.861	.205	.545
COPv After	Extension	-.182	.770	.218	.679	.423	.194
RMSv Before	Extension	-.929	.023*	-.053	.920	.106	.756
RMSv After	Extension	-.681	.206	.123	.817	.265	.431
COM Mean Position Before	Extension	-.967	.007*	-.489	.325	.524	.098†
COM Mean Position After	Extension	.911	.031*	-.785	.065†	.336	.312
COP Min Amplitude	Extension	.081	.897	.064	.905	-.071	.835
COM Min Amplitude	Extension	.289	.637	-.469	.348	.214	.528
COP-COM Min Amplitude	Extension	.554	.332	-.271	.603	.343	.302
COP-COM Max Amplitude	Extension	.326	.674	.224	.669	.078	.829
COP Min Latency	Extension	-.967	.007*	-.541	.267	-.852	.001*
COM Min Latency	Extension	-.466	.428	.459	.359	-.349	.293
COP-COM Min Latency	Extension	-.127	.839	-.937	.006*	.006	.986
COP-COM Max Latency	Extension	.515	.485	.855	.030*	.491	.150
COP Onset	Extension	-.125	.842	.468	.350	.401	.222
COM Onset	Extension	.370	.540	.087	.871	.112	.743
COP-COM Onset	Extension	-.350	.564	-.045	.933	-.290	.387
COP Offset	Extension	-.325	.594	.350	.496	.050	.883
COM Offset	Extension	-.774	.124	.204	.698	-.002	.995
COP-COM Offset	Extension	-.538	.350	.348	.499	-.443	.172

Table A.3.c. Correlations between strength at the ankle in dorsiflexion & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength - Ankle	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Dorsiflexion	-.485	.407	.766	.131	.323	.362
RMSd After	Dorsiflexion	.550	.337	.630	.255	.512	.131
COPv Before	Dorsiflexion	-.448	.449	.628	.257	.297	.405
COPv After	Dorsiflexion	.160	.797	.364	.547	.298	.403
RMSv Before	Dorsiflexion	-.833	.080†	.817	.091†	.433	.211
RMSv After	Dorsiflexion	.187	.764	.594	.291	.419	.228
COM Mean Position Before	Dorsiflexion	.838	.076†	.448	.450	.663	.037*
COM Mean Position After	Dorsiflexion	.822	.087†	.607	.278	.704	.023*
COP Min Amplitude	Dorsiflexion	.461	.434	.504	.387	.441	.202
COM Min Amplitude	Dorsiflexion	.536	.352	.663	.223	.571	.085†
COP-COM Min Amplitude	Dorsiflexion	-.011	.986	-.709	.180	-.130	.721
COP-COM Max Amplitude	Dorsiflexion	.519	.481	-.863	.060†	-.428	.250
COP Min Latency	Dorsiflexion	-.515	.375	.838	.076†	-.018	.962
COM Min Latency	Dorsiflexion	.377	.531	.411	.491	.260	.468
COP-COM Min Latency	Dorsiflexion	-.915	.029*	.414	.488	-.237	.511
COP-COM Max Latency	Dorsiflexion	-.686	.314	-.418	.484	-.414	.268
COP Onset	Dorsiflexion	.680	.206	.421	.480	.528	.117
COM Onset	Dorsiflexion	.319	.600	-.465	.429	-.027	.940
COP-COM Onset	Dorsiflexion	.468	.426	-.127	.838	.107	.769
COP Offset	Dorsiflexion	-.191	.758	.052	.934	-.083	.820
COM Offset	Dorsiflexion	.488	.405	.318	.602	.381	.277
COP-COM Offset	Dorsiflexion	.006	.992	-.222	.719	-.110	.762

Table A.3.d. Correlations between strength at the ankle in plantarflexion & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength - Ankle	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	Plantarflexion	-.217	.726	.226	.714	.128	.724
RMSd After	Plantarflexion	.234	.704	-.123	.844	.155	.669
COPv Before	Plantarflexion	-.030	.961	.137	.826	.190	.599
COPv After	Plantarflexion	.724	.167	.007	.991	.325	.359
RMSv Before	Plantarflexion	-.486	.407	.221	.721	.190	.599
RMSv After	Plantarflexion	.691	.197	.061	.922	.319	.369
COM Mean Position Before	Plantarflexion	.344	.571	.725	.166	.571	.084†
COM Mean Position After	Plantarflexion	.294	.631	.708	.181	.443	.200
COP Min Amplitude	Plantarflexion	-.107	.865	.130	.835	-.062	.865
COM Min Amplitude	Plantarflexion	.025	.968	.686	.201	.268	.454
COP-COM Min Amplitude	Plantarflexion	.295	.630	-.060	.923	.204	.571
COP-COM Max Amplitude	Plantarflexion	-.327	.673	-.088	.888	-.201	.604
COP Min Latency	Plantarflexion	-.433	.467	-.025	.968	-.350	.321
COM Min Latency	Plantarflexion	.013	.983	-.587	.298	-.225	.532
COP-COM Min Latency	Plantarflexion	-.552	.335	.593	.292	.132	.715
COP-COM Max Latency	Plantarflexion	-.016	.984	-.676	.210	-.534	.138
COP Onset	Plantarflexion	.155	.803	-.038	.951	.184	.610
COM Onset	Plantarflexion	-.411	.492	-.736	.156	-.580	.079†
COP-COM Onset	Plantarflexion	.201	.745	-.780	.120	-.400	.253
COP Offset	Plantarflexion	-.462	.434	-.780	.120	-.569	.086†
COM Offset	Plantarflexion	-.037	.953	-.546	.341	-.297	.405
COP-COM Offset	Plantarflexion	-.157	.801	-.863	.059†	-.462	.179

Table A.3.e. Correlations between flexors/extensors (F/E) ratio at the knee & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength – Knee Ratio	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	F/E	.178	.774	.026	.960	.101	.768
RMSd After	F/E	.687	.200	-.038	.944	.247	.465
COPv Before	F/E	-.377	.531	.133	.801	-.001	.999
COPv After	F/E	.157	.801	.299	.565	.226	.504
RMSv Before	F/E	-.276	.653	.130	.806	.060	.860
RMSv After	F/E	.617	.267	.273	.600	.323	.333
COM Mean Position Before	F/E	.136	.827	.363	.480	.248	.463
COM Mean Position After	F/E	-.121	.847	.480	.336	.092	.787
COP Min Amplitude	F/E	.417	.485	-.340	.510	.110	.747
COM Min Amplitude	F/E	.111	.859	.255	.625	.184	.588
COP-COM Min Amplitude	F/E	-.196	.752	.318	.539	-.104	.762
COP-COM Max Amplitude	F/E	.386	.614	-.221	.674	.034	.925
COP Min Latency	F/E	-.060	.924	.213	.685	-.063	.854
COM Min Latency	F/E	.016	.980	-.782	.066†	-.155	.648
COP-COM Min Latency	F/E	-.935	.020*	.642	.169	-.313	.349
COP-COM Max Latency	F/E	-.886	.114	-.857	.029*	-.871	.001*
COP Onset	F/E	.588	.297	.021	.969	.285	.395
COM Onset	F/E	-.004	.995	-.379	.459	-.126	.712
COP-COM Onset	F/E	-.436	.463	-.639	.172	-.457	.158
COP Offset	F/E	-.631	.253	-.803	.054†	-.630	.038*
COM Offset	F/E	.494	.397	-.818	.047*	-.098	.774
COP-COM Offset	F/E	.196	.752	-.929	.007*	-.105	.758

Table A.3.f. Correlations between dorsiflexors/plantarflexors (DF/PF) ratio at the ankle & COP and COM (p<0.05, † trend)*

		Amputees		Controls		All Participants	
Balance Variables	Strength – Ankle Ratio	Pearson Correlation	p-value	Pearson Correlation	p-value	Pearson Correlation	p-value
RMSd Before	DF/PF	-.085	.893	.164	.792	.033	.928
RMSd After	DF/PF	.508	.382	.391	.515	.160	.660
COPv Before	DF/PF	-.607	.278	.127	.839	-.154	.671
COPv After	DF/PF	-.794	.109	.040	.950	-.255	.476
RMSv Before	DF/PF	-.328	.591	.221	.721	.011	.977
RMSv After	DF/PF	-.645	.240	.171	.784	-.132	.716
COM Mean Position Before	DF/PF	.542	.345	-.272	.658	.047	.897
COM Mean Position After	DF/PF	.646	.239	-.124	.842	.253	.480
COP Min Amplitude	DF/PF	.818	.091†	.292	.634	.566	.088†
COM Min Amplitude	DF/PF	.760	.136	-.116	.853	.365	.300
COP-COM Min Amplitude	DF/PF	-.612	.272	-.334	.583	-.470	.170
COP-COM Max Amplitude	DF/PF	.765	.235	-.418	.484	.021	.957
COP Min Latency	DF/PF	.097	.877	.629	.256	.388	.268
COM Min Latency	DF/PF	.599	.286	.816	.092	.592	.071†
COP-COM Min Latency	DF/PF	-.421	.480	-.194	.754	-.382	.276
COP-COM Max Latency	DF/PF	-.477	.523	.343	.572	.228	.555
COP Onset	DF/PF	.736	.156	.141	.822	.214	.553
COM Onset	DF/PF	.850	.068†	.403	.501	.669	.034*
COP-COM Onset	DF/PF	.407	.497	.784	.116	.640	.046*
COP Offset	DF/PF	.551	.336	.856	.064†	.670	.034*
COM Offset	DF/PF	.655	.230	.830	.082†	.712	.021*
COP-COM Offset	DF/PF	.443	.455	.766	.131	.567	.087†

APPENDIX B: Custom-Made Equipment

Figure B.1. Custom-made proprioception machine.

Figure B.2. Custom-made perturbation platform with two force plates mounted on top. Participants stood on one of the force plates during balance testing.

APPENDIX C: Humac Set-up

Figure C.1. Humac set-up to test concentric isokinetic knee strength in flexion and extension.

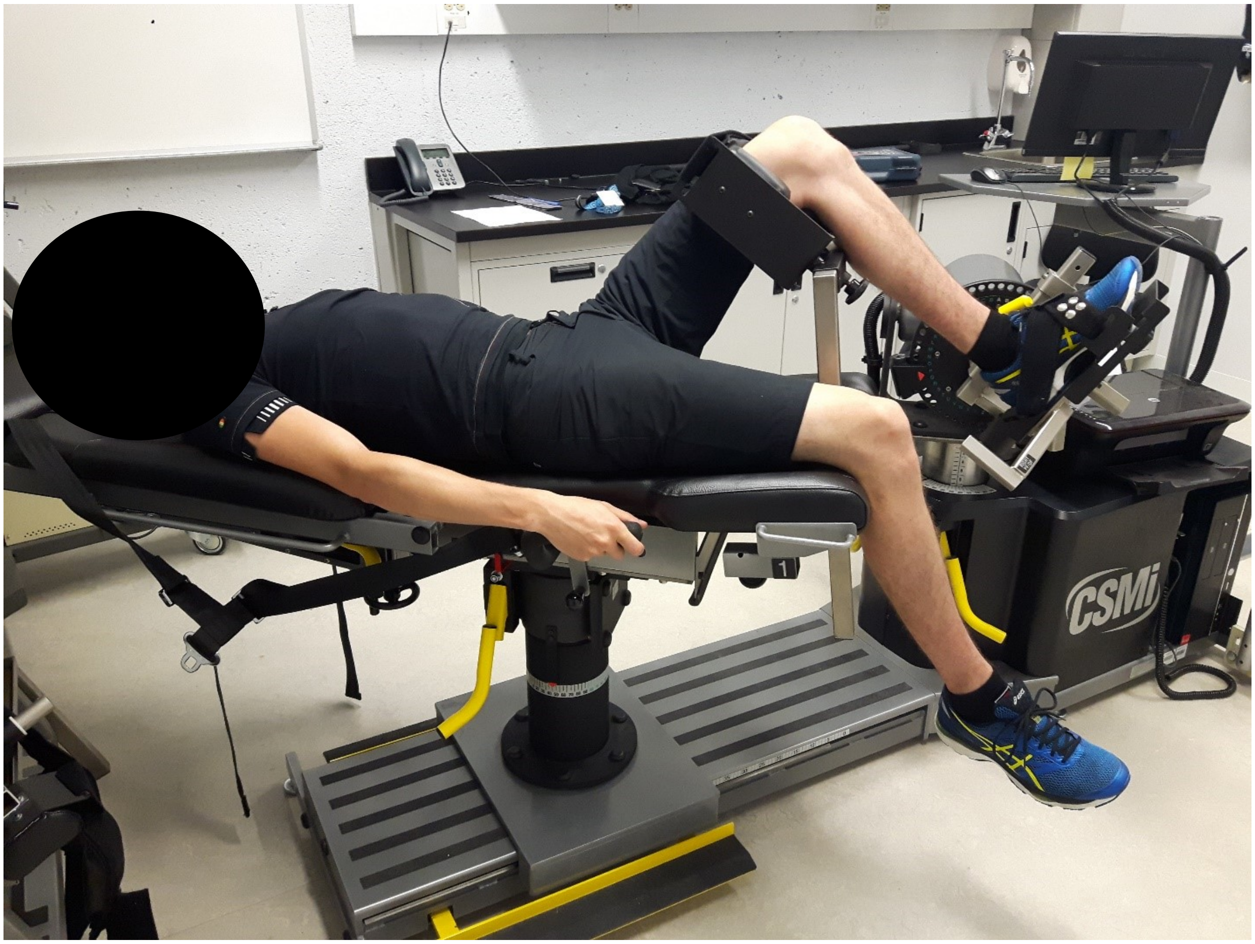

Figure C.2. Humac set-up to test concentric isokinetic ankle strength in plantarflexion and dorsiflexion.