# CHANGES IN SPINAL POSTURE, MUSCLE ACTIVATION, CENTER OF PRESSURE AND DISCOMFORT WHILE STANDING WITH DIFFERENT FOOTREST HEIGHTS DURING A STANDARDIZED COMPUTER TASK

By

Andrew C. Cregg

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Faculty of Health Sciences
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Submitted by: Andrew Cregg

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An oral defense of this thesis took place on November 28th, 2018 in front of the following examining committee:

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| Chair of Examining Committee | Dr. Meghann Lloyd |
| Research Supervisor | Dr. Lori Livingston |
| Examining Committee Member | Dr. Nicholas La Delfa |
| Examining Committee Member | Dr. Paul Yielder |
| External Examiner | Dr. Samuel Howarth, CMCC |

The above committee determined that the thesis is acceptable in form and content and that a satisfactory knowledge of the field covered by the thesis was demonstrated by the candidate during an oral examination. A signed copy of the Certificate of Approval is available from the School of Graduate and Postdoctoral Studies.

**ABSTRACT**

Standing aids are recommended for individuals who are expected to stand for prolonged periods of time in the workplace to reduce the risk of developing low back pain (LBP). The purpose of this study was to examine how footrest usage, and footrests of differing heights, affect measures of posture, muscle activation, weight distribution, centre of pressure, and discomfort while working at a standing workstation. Four standing positions were compared: flat ground stance, and standing with a low (10 cm), medium (20 cm) and tall footrest (30 cm). Using a footrest significantly altered lumbo-sacral angle, lumbar-to-thigh angles, gluteus medius and lumbar erector spinae muscle activity, and $COP_{range}$ in the elevated limb. Discomfort increased over the 15 minute trial regardless of condition. The medium (20cm) footrest provided unique advantages over the other footrest heights and should therefore be recommended in situations where an absolute footrest height is preferred.

**Keywords:** Ergonomics; Footrest; Standing Aid; Intervention; Low Back Pain

## AUTHOR'S DECLARATION

I hereby declare that this thesis consists of original work of which I have authored. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

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The research work in this thesis that was performed in compliance with the regulations of UOIT's Research Ethics Board/Animal Care Committee under REB Certificate number 14477 on August 14th, 2017.

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Andrew Cregg

## STATEMENT OF CONTRIBUTIONS

The following work was conducted in the Occupational Neuromechanics and Ergonomics Laboratory at the University of Ontario Institute of Technology (UOIT). I was responsible for collecting and analyzing all data with the help of a number of individuals. Dr. Lori Livingston, Dr. Nick La Delfa, Dr. Paul Yielder and Ryan Foley contributed significantly towards the design, methods and data analysis portion of the study. Daniel Abdel-Malek, Matthew Russel and Mia Stoicescu contributed towards the data collection process.

I hereby certify that I am the sole author of this thesis and that no part of this thesis has been published or submitted for publication. I have used standard referencing practices to acknowledge ideas, research techniques, or other materials that belong to others. Furthermore, I hereby certify that I am the sole source of the creative works and/or inventive knowledge described in this thesis.

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Andrew Cregg

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# CHAPTER 1: INTRODUCTION, RESEARCH QUESTION AND HYPOTHESES

**Andrew C. Cregg & Lori A. Livingston**

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

Corresponding Author

Lori A. Livingston, PhD

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

2000 Simcoe St. N.

Oshawa, ON, Canada

L1H 7K4

Telephone: (905)721-8668

Email: Lori.Livingston@uoit.ca

Low back pain (LBP) is one of the most common occupational health complaints affecting 84% of the worldwide working population at some point in their lives (Airaksinen et al., 2006). It has been estimated that at any given point in time, 21 million American's suffer from low back pain (Jensen et al., 1994) and over 700,000 worker's compensation claims are submitted per year in the United States of America as a result of occupational LBP (Waddell, 2004). In short, LBP has become a major burden to employers, employees, and other stakeholders affected by lost productivity.

Low back pain has traditionally been associated with manual labour occupations that require repetitive lifting, awkward postures, and vibration (Driscoll, 2014). However, the global working demographic has slowly changed over the past 50 years away from these types of jobs towards those that are office based and sedentary in nature. In fact, the number of moderately physical jobs has decreased by 30% over the past 50 years while the number of light and sedentary jobs has increased to a point where they now represent the vast majority of occupations (Church et al., 2011). Despite this shift, LBP still persists in working populations and sedentary time has now been identified as a modifiable risk factor that can impact LBP (Burton, 2005).

Research has indicated that both sitting and standing can lead to the development of a number of musculoskeletal complaints (Callaghan & McGill, 2001). Sitting places stress on the passive support tissues whereas standing stresses the active support tissues in the back (Callaghan & McGill, 2001). In an effort to find easily administered, cost-effective interventions to reduce sedentary time and LBP in working populations, ergonomists have recommended the use of sit-stand desks. These were developed to provide workers the ability to transfer between standing and sitting throughout the workday, allowing them to take advantage of various spinal postures and cyclically rest tissues to reduce the potential for viscoelastic creep (Callaghan & McGill 2001; Le and Marras, 2016; Mandal, 1981; Nachemson, 1966).

To further reduce low back discomfort, the Canadian Centre for Occupational Health and Safety (CCOHS) and the United States Occupational Safety and Health

Administration (OSHA) have recommended the use of footrests while standing to reduce low back discomfort (Fewster et al., 2017). However, limited research exists to support or refute these recommendations. The purpose of this research is to investigate whether footrest height affects lumbar spine posture, muscle activation, postural stability and discomfort. The inspiration for this research, moreover, came from a collection of studies that investigated standing with a staggered stance, standing on a sloped surface (Gallagher, 2014; Gallagher & Callaghan, 2016), resting one foot on a bar rail, or the use of a footrest (Dolan et al., 1988; Fewster et al., 2017). It is anticipated that this study will add to our knowledge of how footrests may be used to positively impact the health of our working population by reducing the incidence of occupationally induced LBP.

**Research Question**

For healthy young male and female adults, does the height of a footrest affect measures of lumbar spinal posture, muscle activation patterns, centre of pressure and discomfort in comparison to flat ground stance while completing a standardized computer task at a standing desk?

**Hypotheses**

The following hypotheses were tested where $H_o$ represents the null hypothesis and $H_{Alt(n)}$ represents an alternative hypothesis:

*Does gender affect spinal posture, muscle activation or COP?*

$H_o$: Female = Male
$H_{Alt1}$: Female ≠ Male

*Does footrest height affect comfort, spinal posture, muscle activation or COP?*

$H_o$: 10 cm Footrest = 20 cm Footrest = 30 cm Footrest = Flat Ground Stance
$H_{Alt1}$: 10 cm Footrest = 20 cm Footrest = 30 cm Footrest ≠ Flat Ground Stance
$H_{Alt2}$: 10 cm Footrest = 20 cm Footrest ≠ 30 cm Footrest = Flat Ground Stance
$H_{Alt3}$: 10 cm Footrest ≠ 20 cm Footrest = 30 cm Footrest = Flat Ground Stance

$H_{Alt4}$: 10 cm Footrest = 20 cm Footrest ≠ 30 cm Footrest ≠ Flat Ground Stance
$H_{Alt5}$: 10 cm Footrest ≠ 20 cm Footrest = 30 cm Footrest ≠ Flat Ground Stance
$H_{Alt6}$: 10 cm Footrest ≠ 20 cm Footrest ≠ 30 cm Footrest = Flat Ground Stance
$H_{Alt7}$: 10 cm Footrest ≠ 20 cm Footrest ≠ 30 cm Footrest ≠ Flat Ground Stance

*Does time affect comfort, spinal posture, muscle activation or COP?*

$H_o$: Time 5$^{th}$ minute = Time 10$^{th}$ minute = Time 15$^{th}$ minute
$H_{Alt1}$: Time 5$^{th}$ minute = Time 10$^{th}$ minute ≠ Time 15$^{th}$ minute
$H_{Alt2}$: Time 5$^{th}$ minute ≠ Time 10$^{th}$ minute = Time 15$^{th}$ minute
$H_{Alt3}$: Time 5$^{th}$ minute ≠ Time 10$^{th}$ minute ≠ Time 15$^{th}$ minute

*Are there any interaction effects of gender, footrest height or time?*

$H_o$: There are no two-way or three-way interaction effects.
$H_{Alt1}$: There are two-way or three-way interaction effects.

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# CHAPTER 2: OCCUPATIONAL STANDING, LIMB DOMINANCE, AND STANDING AIDS: A REVIEW OF THE LITERATURE

**Andrew C. Cregg & Lori A Livingston**

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

Corresponding Author

Lori A. Livingston, PhD

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

2000 Simcoe St. N.

Oshawa, ON, Canada

L1H 7K4

Telephone: (905)721-8668

Email: Lori.Livingston@uoit.ca

**Abstract**

Occupational health and safety organizations often recommend the use of a footrest during prolonged standing to reduce the risk of developing low back pain (LBP). However, limited research exists to support or refute this claim and even less research attempts to describe the most appropriate footrest specifications. Important factors such as footrest height may have a profound impact on our bodies from a biomechanical perspective. The purpose of this paper is to present a review of the current literature on occupational standing, limb dominance, and standing aids with a particular emphasis on footrests which are the most well accepted standing aid by current research standards. This paper will also review key issues such as occupational LBP, lumbar posture while sitting and standing, muscle fatigue theories, balance, limb dominance, and lateralization concepts. Providing a detailed description of these factors will assist ergonomists and industry leaders in making decisions about workplace design and implementation procedures.

**Key Words:**

Footrest, low back pain (LBP), ergonomics, standing desk, sit-stand desk

**Statement of Authorship:**

Andrew C. Cregg (Masters of Health Science Candidate) was the primary investigator for this investigation and handled all aspects of the research agenda.

Low back pain (LBP) has been identified as the leading cause of disease and years lived with disability worldwide (Driscoll et al., 2014). It affects five out of every ten Canadians within a given six-month period and 31 million Americans at any given point in time (Cassidy et al., 1998, Jensen et al., 1994). Over 700,000 worker's compensation claims are submitted each year for work-related LBP in the United States alone (Waddell, 2004) and is a major contributor towards global disability and mortality (Vos et al., 2015).

The causes of LBP are multifactorial and include physical, social, mental and emotional contributions (Driscoll et al., 2014). To capture the breadth of possible presentations, the definition of LBP has remained vague but regionally specific. It is defined as back pain that lasts for at least one day along the posterior aspect of the body from the 12$^{th}$ rib to the lower gluteal folds (Driscoll et al., 2014). Temporally, LBP can be classified as acute (less than 6 weeks), subacute (less than 12 weeks) or chronic (longer than 12 weeks) which has implications on outcome and prognosis. From an etiological perspective, LBP is broadly classified into two separate and distinct categories: specific and non-specific LBP (Koes et al., 2006). Specific LBP is considered non-mechanical in nature and has an identifiable pathophysiological mechanism such as an infection, disc herniation, osteoporosis, tumor, or fracture. Conversely, non-specific LBP is considered mechanical and does not have a clear cause (Koes et al., 2006). It is often a diagnosis of exclusion and can include any combination of fascial irritation, muscular strains, ligamentous sprains, capsular impingement and joint irritation.

Non-specific LBP represents about 90% of all patients with LBP (Koes et al., 2006) and affects 84% of workers at some point in their lives (Airaksinen et al., 2006; Cassidy et al., 1998). Fortunately, non-specific LBP responds well to conservative interventions such as lifestyle modification and physical therapy (Driscoll et al., 2015). The most common occupational causes have traditionally been associated with prolonged exposure to lifting, forceful movements, awkward postures, and vibration. Males between the ages of 35-66 years who work in agricultural and labour intensive occupations are traditionally considered high risk individuals (Driscoll et al., 2014). However, a subpopulation of workers prone to experiencing LBP has emerged over the past 10 years including workers

who are exposed to prolonged standing periods under poor working conditions. Surveys have shown that 62% of Australian workers (Safe Work Australia, 2011) and 58% of Canadian workers (Tissot et al., 2005) stand for prolonged periods in the workplace (Lee et al., 2017). The emergence of this subpopulation can be attributed to vast changes in the nature of employment over the past 50 years away from manual labour and towards sedentary work environments. In this time, the number of moderately physical jobs has decreased from representing 50% of the workforce to fewer than 20% (Church et al., 2011). Furthermore, light and sedentary jobs now represent the vast majority of occupations in industrialized countries (Church et al., 2011). Public health authorities are justifiably concerned given the strong association between sedentary time and preventable chronic health conditions such as obesity, diabetes, cardiovascular disease (Matthews et al., 2008), and musculoskeletal conditions including low back pain (Callaghan & McGill, 2001; Frymoyer et al. 1980; Nelson-Wong & Callaghan, 2010a).

Sedentary time has been identified as a modifiable risk factor associated with LBP that is both cost-effective and theoretically easy to modify in the workplace (Burton, 2005). A common workplace recommendation is to incorporate standing postures to counteract the effects of sitting. However, prolonged standing is also associated with negative health outcomes such as chronic venous insufficiency (Waters & Dick, 2015) and musculoskeletal pain including LBP (Anderson et al., 2007; Nelson-Wong and Callaghan, 2010b). Between 40-64% of individuals who have no previous history of LBP will develop LBP as a result of prolonged standing (Nelson-Wong & Callaghan, 2010b) and prolonged standing is one of the strongest predictors of LBP (Andersen et al., 2007). As a result, sit-stand desks may be a better recommendation because they allow workers to utilize both sitting and standing postures intermittently throughout the day. Sit-stand desks can reduce prolonged sitting time, increase standing time, and provide intermittent rest for the supporting soft tissues (Callaghan & McGill, 2001; Karol & Robertson, 2015). Previous research suggests that standing desks may be a cost-effective, biomechanically simple way of reducing overall sedentary time (Babski-Reeves & Calhoun, 2016). Understanding the impact that sit-stand desks have in the workplace has become the focus of numerous

research studies over the past 10 years. Widespread implementation of sit-stand desks has been met both with praise and criticism from various groups. To understand how sit-stand desks may impact the incidence and prevalence of LBP, it is important to explore the underlying etiology and common mechanisms of injury associated with such conditions.

## Proposed Mechanisms Underlying LBP

To understand the proposed value of sit-stand desks, one must appreciate the factors that are thought to contribute towards the emergence of LBP. To this end, two lines of research inquiry dominate the literature; that is, understanding the role of spinal posture and muscle activation patterns in the development of LBP.

**Lumbar Spinal Posture.** Posture has become an important clinical indicator for dysfunction and with the emergence of sophisticated and precise measurement tools, it has become the focus of numerous kinematic examinations. Of particular interest to this thesis, previous research has focused on understanding kinetic and kinematic spine characteristics in various sitting and standing positions (Adams & Hutton, 1985; Callaghan & McGill, 2001; Fewster et al., 2017; Gallagher, 2014; Nachemson, 1966; Le & Marras, 2016; Nelson-Wong & Callaghan, 2010a). From a kinematic perspective, sagittal-plane spinal curvature has been shown to have a major impact on the direction and magnitude of forces experienced at the level of the lumbar motion segment (Dolan & Adams, 2001). It has been well documented that humans display stereotypical postures during both standing and sitting to limit caloric expenditure, prevent muscle fatigue and vertebral strain (Augustus et al., 1978; Duval-Beaupere et al., 1992) (Figure 2.1).

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Insert Figure 2.1 Here

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Standing involves lumbar extension (lordosis), hip extension and anterior pelvic tilt (Mandal, 1981). This accentuated lumbar lordosis selectively loads the posterior

annulus and causes compression of the zygapophyseal joints. Conversely, a seated posture is associated with hip flexion, lumbar flexion and posterior pelvic tilt (Mandal, 1981). The moderately flexed posture associated with sitting tends to equalize the compressive forces across the vertebral motion segment and unloads the posterior elements (Dolan & Adams, 2001). However, sitting increases compression to the anterior discs and promotes passive lumbar flexion, high intervertebral disc pressure, and reliance on the passive stabilizing tissues for spinal support (Dolan & Adams, 2001; Nachemson, 1966). Prolonged reliance on the passive stabilizing system can lead to micro-damage accumulation over time, resulting in ligament sprains, capsule irritation, or chronic inflammation (Callaghan and McGill, 2001). Furthermore, sitting without a backrest can increase the intervertebral disc pressure by 40% when compared to standing (Nachemson, 1966). However, this theory has been criticized for ignoring the load bearing capacity of the facet joints (Adams & Hutton, 1980). As such, both sitting and standing postures can lead towards LBP despite stressing different anatomical structures. As a general rule, sitting stresses the passive stabilizing system whereas standing stresses the active stabilizing system and both positions can lead to injury due to fatigue (Callaghan & McGill, 2001).Research has shown that there is no single perfect posture to maintain while sitting or standing and they should therefore be used interchangeably throughout the day to prevent fatigue of either system (Callaghan & McGill, 2001).

If no single optimal posture exists, there may be a variety of postures that can significantly alter the direction and magnitude of forces acting on the spine and ultimately reduce the strain to the low back. One such posture could include having one foot elevated on top of a footrest while standing. This position would theoretically induce slight unilateral hip flexion, posterior pelvic tilt and mild lumbar flexion. Although previous research suggests that deviation from neutral posture can increase the risk of injury (Punnet et al., 1991), workers are not loading their spine while standing in these positions. Additionally, the degree of change is mild and they are not maintaining the position for prolonged periods. Therefore, standing in this position should be acceptable over the short term. Figure 2.2 depicts a proposed postural continuum with standing

situated at one end, and sitting at the other end. Between the two ends of the continuum lies any number of potential intermediary postures.

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Insert Figure 2.2 Here

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Utilization of a footrest would position the body in a posture that falls somewhere along this continuum between sitting and standing in terms of sagittal plane angles. Anecdotally, a low footrest would resemble standing and fall towards the left hand side of the continuum. Alternatively, a tall footrest would fall towards the right side of the continuum and resemble sitting. As the height of the footrest increases from low to high, lumbar posture with become more flexed and progressively resemble a seated posture. That is to say, a low footrest would induce more relative lumbar extension, anterior pelvic tilt and hip extension whereas a high footrest would increase hip flexion, posterior pelvic tilt and lumbar flexion (Bridger et al., 1992; Whistance et al., 1995). Each different footrest height would place the body somewhere within the normal physiological range of motion (ROM) for standing and sitting and may provide an alternative posture for relieving low back discomfort. These variable postures would be accompanied by alterations in muscle activation patterns which may contribute towards reducing fatigue and overall discomfort. A recent study by Glinka and colleagues (2018) examined a similar concept to the postural continuum using a modifiable, custom chair. The chair position participants in various seated postures which transitioned slowly from sitting ($90^o$ of trunk-to-thigh angle flexion) to standing ($180^o$ of trunk-to-thigh angle) in $5^o$ increments. As the chair height increased, participants shifted their body weight from the seat pan to their feet, increased knee extensor activity and increased the pressure between their feet and the floor. The study concluded that chairs that accommodate these intermediate sit-stand postures introduce important tradeoffs between physiologic demand and postural alterations that may have significant impact on comfort and dysfunction. Practitioners should remain mindful of unintended consequences associated with making such recommendations (Glinka et al., 2018).

**Muscle Activation and Fatigue.** Muscle fatigue is defined as the temporary decline in the force and power capacity of skeletal muscle resulting from muscle activity (Potvin & Fuglevand, 2017). Fatigue can manifest peripherally as cross-bridge dysfunction or excitation-contraction coupling impairments and centrally as impaired motor neuron activation (Enoka & Stuart, 1992; Potvin & Fuglevand, 2017). Levels of prolonged muscle activation over 1-2% maximum voluntary contraction (MVC) have been shown to cause muscular fatigue and can be a predisposing factor for LBP (Veiersted et al., 1990). The Cinderella fiber hypothesis suggests that as smaller motor units fatigue, larger motor units begin to fire which increases spinal compression and shear forces (H€agg,1991; Le & Marras, 2016). As prime movers and stabilizers fatigue, antagonist muscles co-activate to maintain stability which imposes different directional loads on the spine which can induce shear and compression in unnatural directions and cause a task to become unacceptable. These novel loads can lead to irregularly higher compressive and shear forces through the intervertebral discs which should not be maintained long term (Veiersted et al., 1990).

As indicated previously, prolonged static loads of two hours could result in injury due to fatigue during both sitting and standing conditions (Callaghan & McGill, 2001). Callaghan and McGill (2001) examined lumbar spine kinematics, spinal loads, and trunk muscle activity while participants remained sitting or standing for two hours. Two hours represents the longest uninterrupted period of static posture for most office employees before taking a break. They observed that standing induced low level static extensor muscle activation which led to fatigue, while prolonged sitting caused static passive tissue stress to the posterior elements including the stabilizing ligaments, joints, joint capsule and tendons leading to fatigue. Therefore, static loading can lead to muscle fatigue and pain in both sitting and standing positions. Fortunately, sitting and standing demonstrated sufficiently different lumbar postures and muscle activation patterns to constitute a rest break for workers who wish to alternate between postures (Callaghan & McGill, 2001). These findings provide support for the use of sit-stand desks and the adoption of a dynamic postural approach in the workplace (Callaghan & McGill, 2001).

When we consider healthy young adults, between 40-60% of individuals are prone to developing LBP after two hours of prolonged standing despite having no previous episodes of LBP (Gregory & Callaghan, 2010; Nelson-Wong & Callaghan, 2010b). These people have a threefold higher likelihood of seeking clinical care for LBP in the future (Fewster et al., 2017; Nelson-Wong & Callaghan, 2014). Individuals who are prone to developing LBP have been labelled as Pain Developers (PD's). Those who do not develop transient acute LBP during a two hour standing trial are labelled as Non-Pain Developers (NPD's) (Nelson-Wong & Callaghan, 2010). There are five main observable differences between PD's and NPD's as follows: PD's report increased discomfort greater than 10mm on the Visual Analogue Scale (VAS) following a two hour standing period, they display poor frontal plane control, high gluteus medius co-contraction, low gluteus medius resting rates, and poor trunk flexor/extensor co-activation (Nelson-Wong & Callaghan, 2010a; Nelson-Wong & Callaghan, 2010b). Conversely, NPD's do not report increased VAS scores following standing, they maintain proper frontal plane control, display a synergistic, reciprocal gluteus medius firing pattern that allows for cyclical resting of the gluteus medius muscles, and have proper flexion/extension co-activation (Nelson-Wong et al., 2008; Nelson-Wong & Callaghan, 2010a). The most clinically significant factors to support the dichotomous division between PD's and NPD's are their VAS scores and gluteus medius co-activation patterns. Figure 2.3 demonstrates the characteristic co-contraction (left) and reciprocal firing patterns (right) associated with PD's and NPD's.

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Insert Figure 2.3 Here

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PD's have been shown to develop pain after approximately 42 minutes of prolonged standing (Coenen et al., 2017; Nelson-Wong, Gregory, Winter & Callaghan, 2008). VAS scores are a retrospective measure that cannot be used to identify PD's prior to the onset of pain. However, gluteus medius co-contraction is more apparent within the first 30 minutes of standing which makes it a potential predetermining factor and a potential screening tool for otherwise healthy individuals (Nelson-Wong, Gregory, Winter

& Callaghan, 2008). Researcher have been able to identify PD's and NPD's with 74% accuracy by observing only gluteus medius co-activation prior to the onset of subjective pain (sensitivity = 0.87, Specificity = 0.5, +LR 1.74, -LR 0.26) (Nelson-Wong, Gregory, Winter & Callaghan, 2008). Additionally, the between-day repeatability of the gluteus medius testing was excellent with 83% of participants remaining in their initial PD/NPD groups.

## Other Factors to Consider In the Study of Standing Postures

The force of gravity is constantly acting on the human body and although not apparent to the naked eye, the human sensorimotor system is continuously active in an effort to keep the body upright. Hence, any consideration of standing work postures must attempt to account for understand how variations in foot position and limb dominance contribute to spinal posture and muscular fatigue.

**Bipedal Stance, Balance and Stability.** Human bipedal stance requires extensive integration of sensory and motor functions to monitor balance and maintain an upright position (Winter, 1995). Balance describes the body's dynamic ability to prevent falls by maintaining the body's center of mass (COM) within its base of support (BOS) (Winter, 1995). The simple act of quiet, unperturbed stance is a continual process of positional monitoring and muscular readjustments. Seemingly static posture is actually a dynamic process that involves constant and predictable oscillatory motion back and forth in the anterior/posterior (A/P) and medial/lateral (M/L) directions (Wang & Newell, 2014; Winter, 1995). Force platforms provide quantifiable kinetic data in the form of ground reaction forces (GRF's), center-of-pressure (COP) location and displacement measurements in three dimensions. COP refers to the summation of vertical ground reaction forces acting over the surface area in contact with the ground (Winter, 1995). The location of the net center of pressure ($COP_{net}$) within the BOS has long been accepted as the controlling variable for postural sway and balance (McCollum & Leen, 1989; Winter et al., 1993).

The original inverted pendulum model developed by David Winter (1993) describes a bottom up, ankle strategy that controlled sway in the A/P direction and a top-down hip strategy that controlled sway in the M/L direction (Winter, 1993). Winter's (1993) was the first documented scenario where multiple force platforms were used to examine balance under each limb rather than simply examining $COP_{net}$. This once novel approach has become common practice while examining balance and postural stability. Over the years, our understanding of postural control has evolved to explain complex, multi-linkage models that combine ankle and hip strategies during various symmetrical and asymmetrical postures (Horak & Nasher, 1986; Hsu et al., 2007; Wang et al., 2014; Winter, 1993).

Symmetrical stance is defined as loading approximately 50% of an individual's body weight to either limb (Wang et al., 2012) whereas asymmetrical posture is defined as placing 65% of an individual's body weight to one limb and the remainder on the other limb (Gallagher, Nelson-Wong & Callaghan, 2011). Common examples of asymmetrical postures include tandem stance, staggering ones stance and using a single legged footrest. Wang and colleagues (2012) found that individuals load between 65-75% of their body weight on the hind limb during tandem and staggered stance which suggests a major supportive role of this limb (Wang & Newell, 2014). Since one limb is in control of the vast majority of an individual's body weight, it's important to consider how limb dominance and cerebral lateralization influence balance and motor control strategies.

**Limb Dominance and Lateralization.** The concept of cerebral lateralization suggests that certain brain functions occur on one side of the brain over the other. Research has attempted to identify behavioural predictors of cerebral lateralization with varying degrees of success (Elias et al., 1998). Limb dominance has long been considered one such predictor. Limb dominance is influenced by genetics, accidental development, environmental factors, cultural factors, and is generally considered task dependent (Grouios et al., 2009; Sadeghi et al., 2000). Studies tend to focus more closely on the relationship between hand preference and cerebral lateralization rather than foot preference. Hand dominance is easier to study and requires more fine motor tasks which

improves testing sensitivity. Traditionally speaking, people can be right-handed, left-handed, cross-dominant, or ambidextrous.

Although most research focuses on handedness, studies suggest that footedness may provide a more reliable indicator of functional cerebral lateralization. Footedness is less influenced by social pressures, cultural variation, and motor habit influences than handedness (Elias et al., 1998; Grouios et al., 2009). Based upon twin studies, genetics account for approximately 25% of the variance in handedness whereas environmental factors can account for up to 75% of the variance (Medland et al., 2006). Oftentimes children feel social pressure from teachers and peers to complete tasks with the right hand (e.g., handwriting) whereas these pressures are not as common during lower limb tasks. Additionally, most motor tasks are designed for right-handers (e.g. cutting with scissors, using a computer mouse) which further drives preferential right handed development.

Studies have attempted to understand the complex relationship between handedness and footedness with mixed results. A study by Barut and colleagues (2007) investigated the relationship between handedness and footedness among 633 individuals between 18-43 years of age. Seventy-five percent of right-handed males and 89.9% of right-handed females were also right-footed but only 57.9% of left-handed males and 79.4% of females were left-footed. A similar study suggested that 95% of right-handed people kick a ball with their right foot whereas 50% of left-handed people kick a ball with the left (Peters, 1979). Likewise, Dida (1988) found that 92% of right-handed people are also right-footed whereas 51% of left-handed people are left-footed. Clearly, right-handed individuals are often right-footed. However, left-handed individuals are much less consistent. When asked to complete a task requiring individuals to mobilize an object with one foot, right-handed individuals stood on their left leg and used their right foot for mobilization over 90% of the time. Left handed individuals stood on their right leg 60-80% of the time, further highlighting the inconsistent dominance pattern amongst left handers (Beling et al., 1998). These studies demonstrate that most people are ipsilaterally dominant regardless of gender but left-handers are contralaterally dominant more often

than right handers. Despite the mounting research on laterality, a direct linkage between handedness and footedness has not been drawn, and it is likely task dependent and heavily influenced by environmental development (Beling et al., 1998; Elias et al., 1998; Wang et al., 2014).

Anatomical studies that focus on examining cross sections of the human spinal cord reveal neuroanatomical evidence of limb dominance in the central nervous system (CNS). The side of the spinal cord corresponding to the dominant limb has a significantly larger cross sectional area (CSA) as a result of the increased corticospinal tract diameter associated with increased neural supply (Nathan, Smith & Deacon, 1990). Additionally, cerebral cortex synaptic interconnectivity is more complex on the side that corresponds to the dominant limb and the dominant hand is represented by a larger area within the primary motor cortex (Hammond 2002; Hammond & Garvey, 2006; Kalayioglu et al., 2008). Therefore, physical evidence exists to support the notion that humans develop limb lateralization, but it cannot be anatomically examined until post-mortem analysis.

To examine limb laterality in the living, academics have long debated whether dominance is best determined based upon questionnaires, performance tasks, or a combination of both (Corey et al., 2001). Clinicians and researchers prefer questionnaires because they are easier and faster to administer and often result in a dichotomous division of left and right handed individuals (Corey et al., 2001). However, these inventories usually focus on uni-manual actives such as hand writing or throwing a ball while ignoring multi-limb tasks and lower limb activities. Results from performance based, bimanual measures often conflict with the results of questionnaire based inventories (Beling et al., 1998). When assessment includes performance measures, the typical bimodal distribution of left and right handers became less apparent (Corey et al., 2001). When two or more performance measures are used, the dichotomous distribution improves.

Despite the long held acceptance that everyone can be classified as either right or left handed, Peters and Murphy (1992) suggest that many more classifications may exist.

They suggest that at least three, but likely five distinct handedness groups exist. According to Peters and Murphy (1992), an accurate classification system would dissociate proximal and distal limb movements. Since proximal and distal muscles are supplied by different descending neural pathways, they may develop to become differentially dominant and result in crossover lateralization within each limb (Corey et al., 2001). Distal limb movements are responsible for fine motor tasks such as hand writing whereas proximal limb movements are more appropriate for gross motor tasks such as throwing a ball. Through this reasoning, an individual who writes with their right hand and throws with their left would be considered proximally left handed but distally right handed.

An alternative classification system suggests that we should divide motor tasks into skilled and unskilled movements based upon anatomical differences in the descending tracts of the spinal cord (Kalayciglu et al., 2008). Unskilled movements (e.g., spinal stabilization) are carried out by proximal and axial muscles which are supplied by the medial/anterior descending pathways (i.e., vestibulospinal and reticulospinal tracts). Alternatively, skilled movements (e.g., hand writing) are carried out by distal and peripheral muscles which are supplied by the lateral descending pathways (i.e., lateral and anterior corticospinal tracts) (Ghez & Krakauer, 2000; Kalayioglu et al., 2008). Regardless of the classification system that is used, research agrees that the question of limb dominance and laterality is not as clear cut as once believed.

We can observe how laterality manifests itself anatomically through neurological structures, but it can also be identified through examination of osseous and muscular development as well (Ozener, 2012). Wolff's Law states that biomechanical factors such as loading can impact morphological changes which leads to tissue strengthening along specific stress lines (Ozener, 2012). As such, when we examine bone density, we observe reactive changes in the diaphyseal CSA of long bones on the dominant side when compared to the non-dominant limb (Ozener 2012; Krahl, 1994). These studies focus on examining bones of unilateral athletic populations such as tennis players and pitchers who develop extreme lateral preferences for one side of their body. A radiographic study of 20 high ranking tennis players demonstrated increased bone density and diameter in

their striking forearm and hand bones when compared to their opposite hand and to a control group (Krahl, 1994). The difference is more pronounced in extreme right handers and is likely less apparent in extreme left handers because of their necessity to adapt to conditions of a right dominant world (Ozener, 2012).

While alternative classification systems of handedness have focused on skilled versus unskilled movements, footedness classification systems focus on differentiating between the mobilizing and stabilizing limb. During unilateral tasks such as kicking a ball, the dominant foot is generally the mobilizing limb which is used for manipulating objects and the non-dominant limb provides postural stability (Peters, 1988; Schneiders et al., 2010; Wang & Newell 2014). To draw a parallel to hand preference, skilled movements are carried out by the dominant, manipulative limb whereas unskilled movements are completed by the non-dominant, stabilizing limb (Kalayioglu et al., 2008; Schneiders et al., 2010; Wang & Newell, 2014). People are generally less aware of their foot preference than their hand preference because typically both lower limbs move symmetrically during daily activities like walking, standing, and running (Wang et al., 2013). As such, footedness is not as well understood as handedness (Peters 1988, Wang et al., 2014). However we do know that footedness appears to be highly task specific like handedness (Wang et al., 2013). For example, while asked to trace a large circle with their foot while standing on their other foot, individuals with utilize a proximal joint strategy involving the hip. Conversely, when asked to trace a small circle, individuals will utilize a distal joint strategy dominated by the ankle (Wang & Newell, 2014). This pattern corresponds very closely to the handedness where proximal muscles dominate gross motor patterns and distal muscles govern fine motor patterns.

A debate that continues to perplex experts is whether the dominant or non-dominant limb requires more neural drive, and the answer may actually be context specific. One theory suggests that the non-dominant, stabilizing limb requires more neural drive to control the weight of the torso and react adequately to perturbations while providing stable support for smooth and precise movements of the mobilizing limb (Wang & Newell, 2014). The alternative school of thought suggests that the dominant,

mobilizing limb requires higher neurological demand because it requires a larger degree of precision and likely many more small motor units (Sadeghi et al., 2000). Although both theories have sound reasoning, neither has been adequately analyzed and the answer remains unknown.

Understanding the difference between dominant/non-dominant and stabilizing/mobilizing limbs is important to consider when studying workplace standing interventions that utilize asymmetrical positions. Asymmetrical positions force the limbs to assume different motor control strategies, muscle activity patterns and loading profiles. Various standing interventions have been examined in the literature that range from footrests and sloped surfaces to treadmill and stationary bicycle stations. It is important to review previous literature involving standing aids in the workplace to understand where gaps still exist and to gain insight into the breadth of knowledge already accumulated in this field of ergonomics.

## Occupational Standing and Device Aids

Occupational health and safety authorities including the Canadian Center of Occupational Health and Safety and the United States' Occupational Safety & Health Administration (OSHA) suggest the use of standing aids to reduce low back pain during prolonged standing in the workplace (CCOHS, 2008; Fewster et al., 2017; OSHA, 2012;). Ergonomic recommendations should aim to optimize lumbar posture, muscle activation patterns and postural stability while minimizing costs associated with workplace modifications. As such, research should focus on quantifying the advantages and disadvantages to each system while considering workplace applicability and financial commitment associated with implementation. The most commonly recommended standing aids include anti-fatigue mats, staggering one's stance, leaning on the desk, standing on a sloped surface and elevating one foot on a footrest (Damecour et al., 2009; Fewster et al., 2017; Gallagher, 2014; Ebben, 2003; Mohan, 2014). Less common recommendations include the use of an extended thoracic support, perching on a stool, walking at a treadmill desks, or spinning on a stationary bicycles (Shrestha, 2016). Given

that extremely limited evidence exists to support these latter recommendations, they will not be examined further in this review. This section will discuss a variety of commonly used standing aids and finish with a description of our current understanding of footrests. Figure 2.4 demonstrates five of these commonly held standing positions that have been examined in the literature.

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**Common Standing Aids.** Anti-fatigue mats have been proposed as a method of reducing subjective discomfort and leg fatigue during prolonged periods of standing (Rys & Kons, 1994). They work by introducing slight instability that requires variable muscle activation over time to react to micro-perturbations (Rys & Kons, 1994). This intervention is inexpensive, easy to implement and has been successfully implemented widespread across the workforce.

Individuals who stagger their stance are effectively increasing their BOS in the A/P direction which improves their overall stability. In this position, workers can shift their weight in the A/P or M/L direction to introduce variable sway that differs from their typical pattern (Gallagher, 2014; Zacharkow, 1988). However, trials involving staggered stance have failed to show a significant impact on lumbar spine posture or muscle activity when compared to flat ground stance (Fewster et al., 2017).

Leaning on a desk is a common workplace posture that can alleviate upper limb and torso discomfort during prolonged standing. However, Damecour and colleagues (2008) found that leaning on a desk did not impact spinal joint angles or alter muscle activity patterns during supported or unsupported thoracic leaning. It seems that the major benefit of leaning on a desk may be to reduce weight bearing by transferring a portion of weight through the torso and upper limbs. However it does not appear that this benefit transfers from the low back into the lower limbs in the same manner.

Standing on a sloped surface alters ankle dorsiflexion and plantar flexion. These changes in joint angles can follow up the kinetic chain and impact lumbar spine posture and muscle activation patterns. A radiographic study found that standing on a declined sloped surface produced significant L1/2 flexion whereas having one foot on an elevated surface produced lumbosacral flexion (i.e., at L5/S1) (Gallagher, 2014). This suggests that the declined surface causes selective upper lumbar flexion whereas the elevated surface produce selective lower lumbar flexion. Another study examined preference between a declined and incline surface and found that individuals preferred the declined surface significantly more often (Nelson-Wong & Callaghan, 2010b). A decline platform results in greater ankle dorsiflexion, trunk flexion, posterior pelvic tilt and lumbar flexion whereas the inclined surface resulted in greater trunk extension, anterior pelvic tilt and lumbar extension.

**Footrests in the Workplace.** The use of a footrest has been proposed to reduce low back pain by impacting hip, pelvis and lumbar spine posture, decreasing intervertebral disc stress and altering muscle activation patterns (Dolan et al., 1988; Fewster et al., 2017; Son et al., 2017; Whistance et al., 1995; White & Panjabi, 1990). Dolan and colleagues (1988) found that having one leg elevated on a 20 cm platform increased lumbar flexion by six degrees and increased low back muscle activity in comparison to flat ground stance. More recently, Fewster and colleagues (2017) investigated the effect that four different standing aids have on muscle activation and lumbar posture. The conditions included: flat ground stance, staggered stance, standing on a sloped surface and standing with a footrest. The height of the footrest was set to allow for $135^0$ of trunk-to-thigh flexion and each individual elevated the same foot that was forward during staggered stance. Contrary to previous research, the sloped surface and staggered stance conditions did not significantly affect lumbar posture (Fewster et al., 2017; Gallagher et al., 2013, Nelson-Wong & Callaghan, 2010a). Only the footrest condition was successful at inducing relative lumbar flexion compared to flat ground stance. They did not observe any significant changes in muscle activity as a result of using a footrest.

To our knowledge, only one previous study has examined how different footrest heights affect the body from a biomechanical perspective (Son et al., 2017). In this investigation, the authors were interested in examining how a footrest adjusted to 5%, 10% and 15% of an individual's body height affects muscle activity, spinal joint angles, balance and pain intensity in participants with intermittent non-specific LBP during a two hour standing trial (Son et al., 2017). The 10% body height footrest resulted in the least fatigue based upon mean power frequency, lowest external moment in the lumbar spine and the least amount of pain. As such, they concluded that health authorities should recommend a footrest that can be adjusted to accommodate 10% of an individual's body height. However, this study was conducted on individuals with a history of non-specific LBP and cannot be directly applied to a healthy population.

Another important measurement to consider while investigating footrests is to examine balance, postural stability and how weight distribution changes within each limb during asymmetrical postures. Mohan and colleagues (2014) examined how standing with a 15cm footrest affected COP measures when the footrest was placed in front and beside the individual. They found that during flat ground stance, participants equally distributed their body weight between the right and left feet (50%/50% distribution). When a footrest was introduced, the percentage body weight distribution changed significantly such that 80% of the weight was applied to the stance limb and 20% was placed on the elevated limb (Mohan et al., 2014). This 80%/20% distribution was consistent regardless if the footrest was placed in the frontal or sagittal plane. It is unknown, however, if this observed weight distribution pattern is dependent upon the height of the footrest.

When we broaden our scope outside of the biomechanical realm, it becomes apparent that discomfort is an important factor to consider when attempting to optimize worker compliance while introducing a new intervention. From an ecological validity standpoint, it may be more effective to recommend absolute footrest heights instead of relative footrest heights based upon worker body height. Workers are more likely to follow recommendations that tell them exactly which height to use rather than measuring for themselves. Rys and Konz (1989) found that workers are more comfortable while using

a 10 cm platform during a four hour standing trial when compared to flat ground stance. Keegan (1953) suggests that $135^0$ of trunk-thigh angle should be recommended because it maintains the physiological normal joint angle of the lumbar spine and positions the muscles of the thighs in a neutral, balanced length that occurs while side-lying in bed. This trunk-thing angle value has been used for standardization purposes in other investigations (Fewster et al., 2017). Other authors have made recommendations based upon field observation (e.g., bar rails and step stools). The most commonly recommended footrest height is between 10-20 cm or 10% of body height (Ebben, 2003; Son et al., 2017). This recommendation attempts to combine the best evidence to provide an objective recommendation. However, this recommendation reflects the wide variability amongst footrest heights within the research and this variability is paralleled by the wide variety of adjustable footrest designs that are available for purchase. Figure 2.5 illustrates some common design options within the market.

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In conclusion, despite the growing amount of evidence supporting the use of footrests, researchers cannot agree upon an appropriate footrest height that best suits all workers. Perhaps, a “one size fits all” approach does not exist due to anthropometric differences and personal preference. However, formal investigation is warranted to determine if appropriate footrest heights can be identified and if a single, standardized footrest height can satisfy all workers needs to improve compliance and ecological validity.

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***Figure 2.1: Lumbar Spine Posture in Sitting and Standing Positions. Images A-E represent progressive changes in lumbar posture from standing to sitting (Keegan, 1953)***

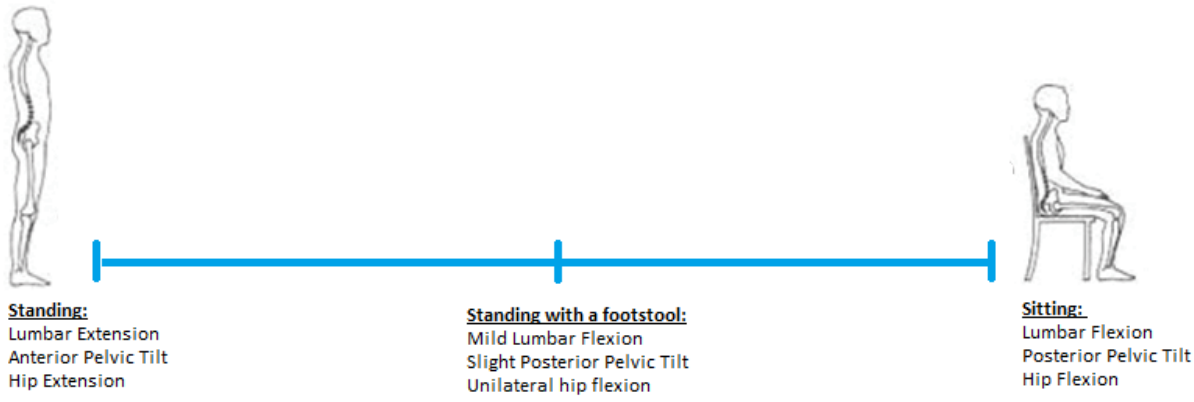



***Figure 2.2: The Proposed Postural Continuum with Standing on the Left End and Sitting on the Right End of the Continuum***

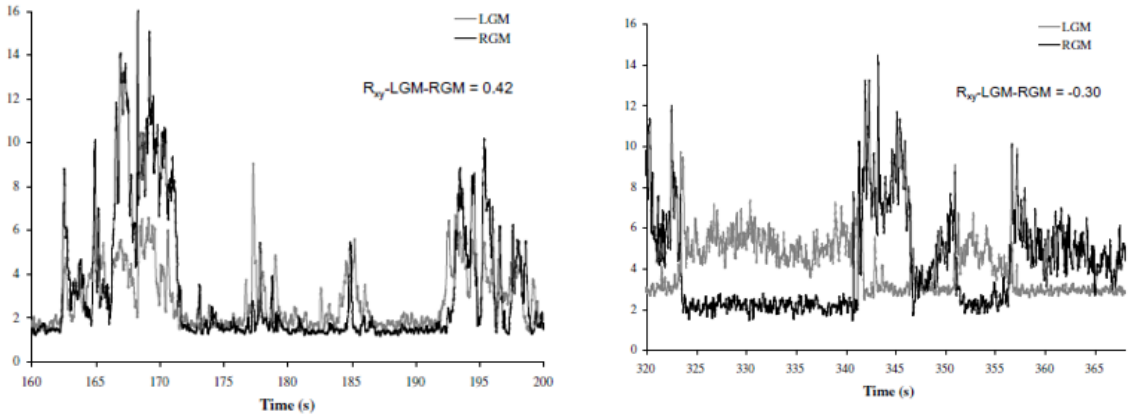
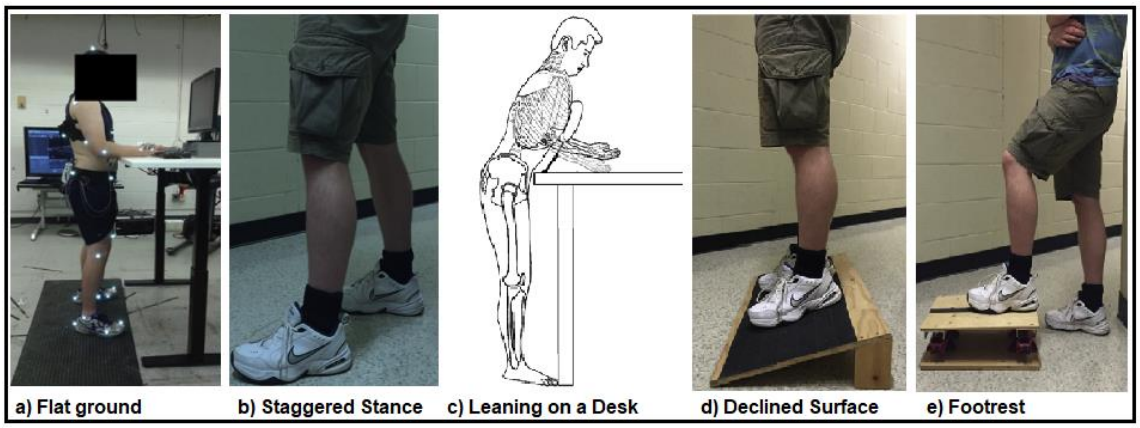



***Figure 2.3: Bilateral gluteus medius EMG activity in Pain Developers (PD's) on the left and Non-Pain Developers (NPD's) on the right (Nelson-Wong, Gregory, Winter & Callaghan, 2008)***




***Figures 2.4 a-e : Commonly studied working postures. (Figure a : Le & Marras, 2016, p.172; Figure c: Damecour et al., 2009, p. 537; Figures 2.4b,d,e: Fewster et al., 2017, p. 283)***

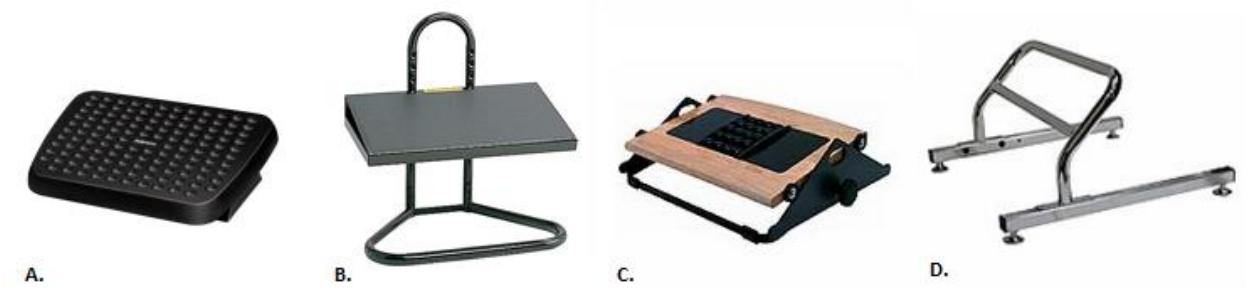



***Figure 2.5a: Followers Standard Footrest, Figure 2.5b: Safco Adjustable Footrest, Figure 5c: Humanscale FM 300B Foot Machine; Figure 2.5d: Stand2Learn Standing Desk Footrest***

# CHAPTER 3: FOOTREST HIEGHT AFFECTS POSTURE AND MUSCLE ACTIVATION DURING SUSTAINED STANDING COMPUTER WORK

**Andrew C. Cregg, Ryan C.A. Foley, Nicolas J. La Delfa & Lori A. Livingston**

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

Corresponding Author

Lori A. Livingston, PhD

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

2000 Simcoe St. N.

Oshawa, ON, Canada

L1H 7K4

Telephone: (905)721-8668

Email: Lori.Livingston@uoit.ca

**Abstract**

Occupationally induced low back pain can result from prolonged static postures including siting and standing. Sit-stand desks are a common workplace recommendation that aims at promoting a dynamic postural approach that allows workers to utilize both sitting and standing postures. For workers who are required to stand for prolonged periods, occupational health and safety experts recommend the use of a standing aid to reduce the incidence of low back pain (LBP). The most commonly recommended standing aid is a single legged footrest. However, only one study has examined how footrest height affects the body from a biomechanical perspective and it was conducted on a clinical population. The purpose of this paper is to examine how utilization of a footrest affects joint positions and muscle activity and to determine if these affects are dependent upon the height of a footrest. Independent factors included kinematic analysis of spinal and hip joint angles and muscle activity patterns including the gluteus medius (GM), lumbar erector spinae (LES) and tensor fascia lata (TFL).

Sixteen healthy young participants worked at a standing workstation for 15 minutes under four conditions: flat ground stance, using a low footrest (10 cm), medium footrest (20 cm) and a high footrest (30 cm). The results of this study suggest that footrest utilization induces flexion through the thoraco-lumbar, lumbo-sacral and lumbar to elevated thigh angles. The degree of change in the lumbar to elevated thigh flexion was offset by a similar degree of extension through the stance limb. When a footrest was present, the GM activity on the elevated limb side was reduced whereas the LES activity was elevated. These patterns correspond to a reduced stability demand through the hip joint and increased stabilization demand through the lumbar spine respectively. As such, utilization of a footrest between 10-30 cm significantly alters posture and muscle activation. Therefore, utilization of a footrest may provide a tertiary posture to be used within a dynamic postural approach in the workplace. The observed effects were not dependent upon footrest height.

**Key Words:**

Footrest, low back pain (LBP), ergonomics, standing desk, electromyography, kinematics, posture

**Statement of Authorship:**

Andrew C. Cregg (Masters of Health Science Candidate) was the primary investigator for this investigation and handled all aspects of the research agenda.

## Introduction

Low back pain (LBP) has been identified as the leading cause of disease and years lived with disability worldwide with a global point prevalence of 18.3% and a 1-month prevalence of 30.8% (Maher, Underwood & Buchbinder, 2016; Vos et al., 2015). Non-specific LBP is the term given to LBP generated by structures in the musculoskeletal system without a specific, pathophysiological cause and it accounts for approximately 90% of all cases (Koes & van Tulder, 2006). Moreover, non-specific LBP has become a major occupational health hazard affecting 84% of all workers at some point during their lifetime (Airaksinen et al., 2006; Cassidy et al., 1998).

Modern day workers spend the vast majority of their day in prolonged, sedentary working postures such as sitting and standing (Church et al., 2011). Prolonged static loads of two hours have been shown to result in injuries including LBP due to muscle fatigue in both the seated and standing position (Callaghan & McGill, 2001). The current evidence suggests that workers should adopt a dynamic postural approach in the workplace, one that takes advantage of routinely shifting between a seated and standing position to best prevent LBP (Gallagher, Campbell & Callaghan, 2014). Sit-stand desks are a common ergonomic intervention that promotes this dynamic postural approach. However, workers continue to experience LBP and ergonomists persist in their search for cost effective and simple to implement interventions to prevent LBP.

Within a population of healthy young individuals, 40-71% will develop transient LBP during prolonged standing periods despite having no previous history of LBP (Gallagher, Nelson-Wong & Callaghan, 2011; Marshall, Patel & Callaghan, 2011; Nelson-Wong & Callaghan, 2010). Individuals generally fall into two broad categories: Pain Developers (PDs) or Non-Pain Developers (NPDs) based on their predilection towards developing pain following prolonged periods of standing (Nelson-Wong & Callaghan, 2010b; Nelson-Wong, Gregory, Winter & Callaghan, 2008). The two most clinically applicable methods for identifying these individuals include administering a pain indicative Visual Analogue Scale (VAS) or analysis of gluteus medius co-activation patterns

bilaterally. PD's will experience a significant change in VAS score (>10 mm) following a two hour standing period where as NPD's will not (Nelson-Wong & Callaghan, 2010a). More importantly for screening purposes, PDs display high gluteus medius co-contraction patterns within the first 30 minutes of standing whereas NPD's display a synergistic, reciprocal gluteus medius firing pattern. Given that most PDs will demonstrate pain after 42 minutes of prolonged standing, observation of gluteus medius activation patterns may provide a screening mechanism for identifying PDs prior to the onset of pain within the first 30 minutes of standing (Nelson-Wong, Gregory, Winter & Callaghan, 2008).

For workers who are required to stand for prolonged periods of time, occupational health and safety authorities recommend the use of a standing aid to reduce the incidence of LBP (Ebben, 2003; Fewster, 2017). Fewster and colleagues (2017) recently examined four commonly utilized standing positions in the workplace including flat ground stance, having one foot elevated on a footrest, staggered stance with feet positioned on a 45$^{o}$ angle and the use of a sloped surface. The use of a footrest was associated with increased spinal flexion but no changes in muscle activation in comparison to flat ground stance. Staggering stance and using a sloped surface were also seen to be ineffective in changing lumbar posture. Although there were no significant changes in muscle activation levels, participants who developed pain did display the previously cited, characteristically high gluteus medius co-activation pattern associated with PD's (Fewster et al., 2017). Other studies have shown that the use of a footrest can affect the lumbar, pelvic and hip joint angles, decrease intervertebral disc stress, alter weight distribution, increase lumbar erector activity and alter muscle co-activation patterns in the gluteus medius muscles (Dolan et al., 1988; Fewster et al., 2017; Son et al., 2017, Whistance et al., 1995; White & Panjabi, 1990).

Only one study to our knowledge has examined how the height of a footrest affects lumbar spine posture while also monitoring muscle activation. Son and colleagues (2017) examined how normalized footrest height of 5%, 10% and 15% of body height affected lumbo-sacral angles, muscle activity and discomfort in workers with a history of LBP. They found that a footrest normalized to 10% of a worker's height maximizes comfort

and minimizes fatigue. Alternative studies have also attempted to develop recommended footrest heights, however they did not directly compare between various heights. For example, Keegan (1953) recommends that a footrest allowing $135^0$ of trunk-to-thigh angle flexion promotes a resting lumbar angle. Similarly, Dolan (1988) suggested that a 20 cm footrest increases lumbar flexion by $6^o$ and elevates lumbar spinal extensor activity overall. Despite the variability in the approach, occupational health and safety organizations generally agree that workers should use a footrest that falls between 10-20 cm to reduce lumbar strain, optimize muscle activation and improve comfort (Ebben, 2003).

There would be value in identifying a simple, singular recommendation pertaining to footrest height and especially so if it led to increased levels of footrest use. Workers and employers are more likely to implement a simple recommendation that fits all workers rather than measuring each worker's height and matching a footrest to those specifications. Additionally, the difference between a 10cm and 20cm footrest has not been directly examined and the current recommendations fail to provide adequate specifications about how to determine the appropriate height within this range. As such, the purpose of this study was to examine how absolute footrest heights affect posture and muscle activation patterns in healthy young individuals in an attempt to provide a single recommendation for workers. Muscle activation patterns of the lumbar erector spinae, gluteus medius, and tensor fascia lata muscles were bilaterally analyzed, as were joint angles including the thoraco-lumbar, lumbo-sacral, and lumbar-to-thigh angles. It was hypothesized *a priori* that the use of a footrest would significantly affect posture and muscle activation patterns in the above mentioned joints and muscle groups. We expect that our observations will manifest as a graded change in joint angles and muscle activation that are dependent upon the height of the footrest. As such, as the footrest height increases, the stance limb will be associated with gradually increasing joint extension and elevated muscle activity whereas the elevated limb is expected to demonstrate gradual flexion and incremental reductions in muscle activity. The spinal joint angles are expected to become more flexed as the height of the footrest increases.

It is anticipated that this work will help to improve guideline development and implementation protocol surrounding the use of footrests in the workplace.

## Methods

This research was approved by the University of Ontario Institute of Technology (UOIT) Research Ethics Board (REB# 14477) and written informed consent was obtained from all participants prior to data collection (Appendix A). Participants were recruited from the UOIT population through email invitation and posters which were displayed throughout campus. Those who agreed to participate completed the Waterloo Handedness and Footedness questionnaires, Oswestry Low Back Disability Index (OBDI), and provided demographic information prior to participation (Appendix B-D). Participants were excluded from the study if they reported a history of LBP that required medical intervention or time off work greater than three days in the past 12 months, employment in a job that required prolonged standing over the past 12 months, had previous lumbar or hip surgery, an inability to stand for two hours, or a clinically significant score on the OBDI (Fewster et al., 2017).

*Participant Sample:*

Based upon a power calculation ($\alpha = 0.05$, $\beta = 0.80$), sixteen healthy individuals (i.e., 8 male, 8 female) were recruited with a mean age of 23.8 +/- 4.3 years, height 172.8 cm +/- 10.0 cm, and weight 69.9 kg, +/- 15.6 kg participated (Table 3.1). The sample size was calculated using the GLIMMPSE (https://glimmpse.samplesizeshop.org/#/) online sample calculation tool. Using an $\alpha = 0.05$ and $\beta = 0.80$, an effect size was derived from data generated by Fewster and colleagues (2017) and a repeated measures approach, a minimal sample size of 16 was determined. The sample size was based upon kinematic measures because the intervention was expected to have the largest effect on posture. Fifteen participants were right side dominant while one participant was left side dominant based upon the Waterloo Handedness and Footedness Questionnaires. All demographic data and outcome measure scores are displayed in Table 3.1. All data collection occurred

during one test session, on one day over a 2.5 hour period. This study was conducted concurrently with another study that assessed center of pressure (COP) and discomfort.

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Insert Table 3.1 Here

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**Instrumentation:**

The placement of electromyographic (EMG) electrodes and kinematic markers mirrored that of standardized laboratory protocols utilized in previous studies (Fewster et al., 2017; Gallagher, 2014; Zipp, 1982). Figure 3.1 depicts the placement of the EMG electrodes and kinematic markers. The collection of EMG and motion capture data were time synchronized and exported through NDI First Principals Software v1.5™ (Northern Digital Inc., Waterloo, Canada) prior to statistical analysis.

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Insert Figure 3.1 Here

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*Electromyography (EMG):*

A Trigno Wireless Surface EMG system (Delsys, Natick, MS, USA) was used to collect muscle activity from the lumbar erector spinae (LES), gluteus medius (GM) and tensor fascia lata (TFL) muscles bilaterally. Prior to electrode placement, the skin was prepared by shaving, gently abrading, and sanitizing the area with rubbing alcohol to maximize skin adherence and minimize electrical impedance. The electrodes were placed superior to the third lumbar vertebral transverse processes bilaterally for the LES, halfway between the most superior aspect of the iliac crest and the greater trochanter for the GM, and one centimeter posterior to, and midway along the line, connecting the anterior superior iliac crest (ASIS) and greater trochanter for the TFL (Fewster, 2017; McGill, 1991; Nelson-Wong et al., 2008; Zipp, 1982). Data were collected using Delsys EMGWorks v4.5.4

software and recorded using NDI First Principals software v1.5™ (Northern Digital Inc., Waterloo, Canada). EMG signals were collected at a frequency of 2000Hz using a 16-bit A/D conversion card. The signals were differentially amplified using a common mode rejection ratio of 80dB at 60Hz, analogue band-pass filtered from 20-450 Hz and gained by a factor of 909V +/- 5%.

*Motion Capture:*

Kinematic data were collected using an Optotrak 3D Investigator Motion Capture System (Northern Digital Inc., Waterloo, ON) and First Principals Software v1.5™ (Northern Digital Inc., Waterloo, Canada). Rigid bodies were placed at the level of T9, L1/L2 and over the sacrum in the sagittal plane and bilaterally at the midpoint between the greater trochanter and lateral femoral condyles to define the thorax, lumbar, sacral and thigh segments respectively. Joint angles of interest included the thoraco-lumbar, thoraco-sacral, lumbo-sacral and lumbar-to-thigh joints. Kinematic data were gathered at a frequency of 50 Hz (registration RMS error = 0.38 mm +/- 0.028mm, RMS Alignment Error = 0.079mm +/- 0.017mm) from a total of 23 active markers. The local and global coordinate systems (GCS) were set up according to the International Society of Biomechanics (ISB) standards (Wu & Cavanaugh, 1995). A digitizing probe, which contained three markers at a known distance from the tip of the tool, was used to define anatomical landmarks in relation to the rigid bodies. Based upon the known distance between the digitized points and the rigid bodies, a static digital model was developed for each participant to calculate the relative joint angles compared to their resting anatomical position. The digitized points included the left and right acromion processes and posterior lateral ninth rib for the thoracic spine, the ninth rib and PSIS's for the lumbar spine, left and right PSIS's and ASIS's for the sacrum, the left and right greater trochanters and medial and lateral femoral condyles for the femurs.

**Experimental Protocol:**

Once participants had been fitted with the necessary equipment, they conducted EMG and kinematic normalization procedures. Participants were required to stand in the anatomical position for 5 seconds to capture static joint angles which were used as a

baseline measure for experimental conditions. Then, resting EMG was collected, followed by maximum voluntary isometric contractions (MVIC's) for each muscle group. MVIC's were obtained for the LES in the Beiring-Sorensen position, whereas the GM and TFL were collected in the side lying position while resisting combined hip abduction and extension for the GM, and hip abduction with slight flexion for the TFL (Fewster et al., 2017; Nelson-Wong et al., 2008). Each muscle was tested twice separated by one minute rest intervals. For the remainder of the study, the peak MVIC activity (regardless of which muscle was being tested) was considered the maximum voluntary contraction (MVC). Once the normalization process was completed, the collection of EMG and kinematic data were time synchronized by the triggering of a single keystroke.

The experiment included four standing conditions held for 15 minutes each which were separated by brief 3-5 minute rest breaks. The conditions included flat ground stance, standing with a low footrest (10cm), medium footrest (20cm), and a high footrest (30cm). Participants always completed flat ground stance first as a baseline measure followed by the footrest trials in a randomized order. Based upon the results of their Waterloo Footedness Questionnaire participants elevated their dominant foot on the footrest. During each trial, participants completed a standardized data entry task at a standing desk. The desk was modified to allow for 5-6 cm of clearance between the participant's wrists and the table when their elbows were placed at 90 $^{o}$ of flexion (Kroemer & Grandjean, 1997). Using instructions adapted from Gallagher (2014), participants were asked to stand in their usual manner for the entire period of time. When a footrest was provided, participants were asked to maintain contact with the ground or footrest at all times with both limbs. They were instructed to avoid leaning on the table and asked to not lift their foot from the footrest. Following completion of the study, participants were asked which footrest the preferred for subjective analysis.

**Data Processing and Analysis**

During each 15 minute data collection trial, the data were clipped into three, one minute windows for statistical analysis. The windows included data from minutes 2.5-3.5,

7.5-8.5 and 12.5-13.5 which were referred to as the “start” “middle” and “end” phases, respectively. These time bins were chosen because participants were required to complete a rating of perceived discomfort (RPD) questionnaire at the 0, 5, 10 and 15 minute mark for a parallel study. The data were collected in a staggered manner to prevent contamination of signals when participants shifted their body to complete the questionnaires as a data preservation technique.

*Electromyography (EMG):*

Raw EMG signals were exported from NDI First Principals into a custom MatLab (V.3) script for processing. The signals were band-pass filtered with a cutoff frequency of 10-500Hz with a dual-pass, second order Butterworth filter (Mohan et al., 2014). The signals were then debiased and full wave rectified. The resulting linear enveloped signal was low-pass filtered at 2 Hz using a dual-pass Butterworth filter and expressed as a percentage of MVC. The filtered signal was then clipped into the three time windows representing the “start” “middle” and “end” of each trial for analysis as described previously. Mean EMG amplitudes were calculated and exported for statistical analysis in SPSS.

*Motion Capture:*

All kinematic data were collected using NDI First Principles software and imported into Visual 3D (V.4) (C-Motion Inc., Germantown, MD) for initial modeling. Digitized anatomical landmarks were identified and rigid bodies were defined to calculate the raw joint angles. A static, hybrid model was used as a baseline measure to build a digital human skeleton for joint angle calculations. In situations where marker data were lost or missing, data interpolation was conducted to bridge the gap between existing data points for a maximum of 20 frames or 0.4 seconds. Joint angles were expressed as the superior segment in relation to the inferior segment and the resultant angles were imported into a custom MatLab script for further processing. The data were low-pass filtered at 2.5 Hz with a dual-pass, fourth-order Butterworth filter (Hoang, 2016; Winter, 2005). Positive angles represent relative flexion and negative results represent relative extension in relation to anatomical position (Hoang, 2016).

*Statistical Analysis:*

All statistical tests were conducted using SPSS (Version 24.0) for Windows 10 (SPSS, Inc., Chicago, IL, USA). Descriptive statistics were calculated for each dependent variable to represent the central tendency and variability within each data set. Pearson correlation coefficients were calculated to determine the extent to which there was shared variance between dependent variables. Significant Pearson correlations ($\alpha = 0.05$, $\beta = 0.8$) were further examined using a multivariate analysis of variance (MANOVA). All other relationships were examined using independent mixed-between, repeated measures analysis of variances (ANOVA's). The independent variables included one between factor (i.e., sex) and two within factors (i.e., four levels of footrest height; three levels of time). The level of significance was set at $\alpha < 0.05$ for all statistical tests. In situations where the data did not meet sphericity, Greenhouse-Geisser corrected values were used to determine degrees of freedom and the adjusted p-values were used. For significant interactions and main effects, pairwise comparisons were made using post-hoc Tukey's HSD tests.

## Results

*Electromyography (EMG):*

Muscle activation levels were on average low throughout all trials (M = 3.0%, SD = +/-3.0%, Range = 1-9% MVC). The means and standard deviations for all six muscles/muscle groups are summarized in Table 3.2 and the results of the statistical analyses may be found in Table 3.3. Comparisons were made to investigate potential interaction and main effects of time, footrest height, and sex for each of the muscle groups studied.

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Insert Tables 3.2 and 3.3 Here

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A significant main effect of footrest condition was observed for the elevated limb GM activity ($F(1,18) = 8.26$, $p < 0.01$, $\omega^2 = 0.37$) (Figure 3.2). As such, the low and high footrest conditions resulted in significantly lower muscle activation than the flat ground trial. The medium footrest demonstrated the same trend but was non-significant. Also, there was a trend towards the elevated limb having lower GM activation than the stance limb, but they were not significantly correlated ($p = 0.7$, $r = 0.51$). The GM activation on the stance limb also trended towards demonstrating higher muscle activation during footrest trials, but the difference was also non-significant ($F(3,42) = 1.27$, $p > 0.05$, $\omega^2 = 0.08$).

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Insert Figure 3.2 Here

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Significant interaction effects of footrest condition and time ($F(4,54) = 3.15$, $p = 0.02$, $\omega^2 = 0.18$) (Figure 3.3) and footrest condition and sex ($F(2,22) = 4.01$, $p = 0.04$, $\omega^2 = 0.22$) (Figure 3.4) were observed for the elevated limb ES. In comparison to flat ground stance, the low footrest resulted in significantly higher activity over the full trial and the medium footrest resulted in higher activity at the middle and end of the trial (Figure 3.3). Comparing between footrests, the low footrest produced significantly higher activity than the medium footrest at the start and end of the trial and more activity than the high footrest at the middle and end of the trial (Figure 3.3). For males, the footrest conditions resulted in significantly higher elevated limb ES activity, however the medium and tall footrest resulted in significantly lower activity than the low footrest. For females, the highest footrest resulted in significantly lower activation than the flat ground condition (Figure 3.4). A reverse trend was seen between males and females where the presence of a footrest increased elevated ES activity in males and reduced activity in females.

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Insert Figure 3.3 and 3.4 Here

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*Motion Capture Kinematics:*

Summary tables demonstrating the relative mean flexion/extension angles and statistical tests ($\alpha = 0.05$) are presented in Table 3.4 and Table 3.5 respectively. Positive vales represent flexion and negative values represent extension relative to anatomical position.

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Insert Tables 3.4 and 3.5 Here

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A significant main effect of footrest condition was observed for multiple angles including the thoraco-sacral ($F(2,26) = 7.18$, $p < 0.01$, $\omega^2 = 0.34$) (Figure 3.5), lumbo-sacral ($F(1, 21) = 11.61$, $p < 0.01$, $\omega^2 = 0.45$) (Figure 3.6), lumbar to stance thigh ($F(2,24) = 16.50$, $p < 0.01$, $\omega^2 = 0.54$) (Figure 3.7) and lumbar to footrest thigh angles ($F(2,25) = 10.27$, $p < 0.01$, $\omega^2 = 0.42$) (Figure 3.7). Thoraco-sacral and lumbo-sacral angles demonstrated significantly more flexion during footrest trials when compared to flat ground stance. Additionally, all footrest trials produced significantly different angles when compared to each other although these were not likely clinically significant. The medium footrest produced the least change in angle whereas the medium footrest produced the largest change in flexion when compared to flat ground (Figure 3.5 and 3.6). When the lumbar-to-thigh angles were examined, flexion in one thigh was offset by extension in the other thigh. Comparisons between all footrest conditions were significantly different with the medium footrest producing the largest deviation from resting in both limbs (Figure 3.7). During footrest trials, the elevated limb was always in flexion and the stance limb was always in extension. An inverse relationship was observed between the two limbs where flexion in one limb was counteracted by extension in the other limb (Figure 3.7 and 3.8).

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Insert Figures 3.5 to 3.8 Here

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A main effect of time was observed for the lumbo-sacral angle ($F(2,24) = 7.76$, $p < 0.01$, $\omega 2 = 0.36$) (Figure 3.9). As such, the lumbo-sacral flexion angle increased as time progressed regardless of footrest condition or sex.

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Insert Figure 3.9 Here

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A significant main effect of sex was observed for the lumbar to footrest thigh angle ($F(1, 14) = 8.69$, $p = 0.01$, $\omega^2 = 0.38$) (Figure 3.9). Regardless of time or condition, males stood with more overall flexion through the lumbar to footrest thigh angle and females stood with more relative extension.

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Insert Figure 3.10 Here

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Interaction effects of footrest condition and time were observed for thoraco-sacral ($F(3,47) = 3.17$, $p = 0.03$, $\omega^2 = 0.19$) (Figure 3.11), lumbo-sacral ($F(3,46) = 2.83$, $p < 0.05$, $\omega^2 = 0.17$) (Figure 3.12) and lumbar to stance thigh angles ($F(3,42) = 3.03$, $p = 0.04$, $\omega^2 = 0.18$) (Figure 3.13). The thoraco-sacral angle was more flexed during the middle of the low footrest trial and during the middle and end of the medium footrest trial when compared to the tallest footrest (Figure 3.11). The lumbo-sacral angle was significantly more flexed during the start of the low footrest trial and during the end of the medium footrest trial when compared to the high footrest (Figure 3.12). The lumbar to stance thigh angle was significantly less extended during the middle of the low footrest trial and

during the end of the highest footrest trial when compared to the medium footrest (Figure 3.13).

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Insert Figures 3.11 – 3.13 Here

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An interaction effect of footrest condition and sex was observed for the lumbar to stance thigh angle ($F(2,24) = 3.76$, $p = 0.04$, $\omega^2 = 0.21$) (Figure 3.14). For both males and females, footrest trials resulted in significantly more extension than the flat ground trial. In males, the medium footrest resulted in the most extension whereas females had similar extension throughout all footrest trials.

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Insert Figure 3.14 Here

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A significant three way interaction was observed between footrest condition, time and sex for the lumbar to stance thigh angle ($F(3,42) = 3.09$, $p = 0.04$, $\omega^2 = 0.18$) (Figure 3.15). For males, the medium footrest resulted in more extension than the low footrest over the entire trial and more extension than the high footrest towards the end. The start of the low trial was significantly more flexed than during the start of the high footrest condition. For females, the medium footrest resulted in significantly more extension than the high footrest during the middle portion of the trial. The flat ground trial resulted in significantly more flexion than any of the other conditions in both males and females.

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Insert Figure 3.15 Here

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## Discussion

The purpose of this study was to examine how various footrest heights affect muscle activation and joint position while working at a standing desk. Consistent with our hypothesis, the use of a footrest significantly affected both muscle activation and joint position. However, the observed changes in muscle activity were only significant for the elevated limb despite the stance limb showing a trend towards significance. Additionally, the majority of observed effects were not dependent upon footrest height as we expected.

In the elevated limb, the presence of a footrest resulted in reduced gluteus medius activity and elevated lumbar erector spinae activity. Research has shown that individuals place approximately 50% of their weight on either limb while standing on flat ground (Chapter 4; Mohan et al., 2014). In contrast, they place approximately 80-85% of their weight on the stance limb and 15-20% on the elevated limb when a footrest is present, regardless of footrest height (Cregg et al., 2018; Mohan et al., 2014). Therefore when using a footrest, the stance limb is responsible for stabilizing the majority of an individual's body weight and the elevated limb only manages a small fraction of the body weight. When less weight is placed through the elevated limb when using a footrest, there is less need for hip stabilization. Hence the reduction in gluteus medius activation.

Conversely, the erector spinae activity on the side of the elevated limb increased while individuals used a footrest. This reflects the unilateral increase in core stabilization demand associated with the elevated limb during semi-closed chain activities. The core refers to a collection of muscles that increase intra-abdominal pressure to create a rigid cylinder that supports the lumbar spine and pelvis during body movements (Kibler et al., 2006). By stabilizing the lumbar spine, the core acts to provide "proximal stability for distal mobility" to allow the lower limb to perform open chained movements such as stepping, walking and running (Kibler et al., 2006). When one foot is elevated on a footrest, an increase in unilateral erector spinae activity provides lumbo-pelvic frontal plane stability. This creates a solid structural foundation that can support the elevated limb which is acting in a semi-open chained manner with the footrest. The elevated limb demonstrates less contact force with the ground, resulting in a semi-open chained environment that requires increased core stability. To summarize these findings, the observed decrease in gluteus medius activation reflects the reduced need for local stabilization in the hip. In contrast, the increased erector spinae activity reflects the increased need for global core

stabilization through the lumbar spine to support the elevated limb which has reduced contact force with the footrest and acts in a semi-open chained manner.

Sustained muscle activity greater than 1-2% MVC has been shown to significantly contribute towards muscle fatigue (Aaras, 1994; Callaghan & McGill, 2001; Veiersted, 1994). The average muscle activation in the current study was 3% MVC overall. However, the erector spinae associated with the stance limb displayed elevated yet non-significant MVC levels above the recommended 2% threshold. If this level was sustained for a prolonged period of time, it could contribute significantly towards fatigue and injury (Callaghan & McGill, 2001).

Muscle activation changes were accompanied by a number of significant joint angle changes. Most notably, lumbo-sacral flexion increased when a footrest was present (Figure 3.5). These findings are consistent with previous research which found that the use of a footrest influenced participants to stand with a more flexed posture (Dolan et al., 1988; Fewster et al., 2017; Gallagher, 2014). However, contrary to our expectation that spinal flexion would increase as footrest height increased, the low and medium footrest demonstrated a graded effect but the highest footrest did not. The same pattern was observed in the lumbar to thigh angle on the elevated limb side (Figure 3.6). In both the lumbo-pelvic and lumbar to thigh angles, the tallest footrest resulted in the least lumbo-sacral flexion of all the footrests. However, it is important to note that this analysis only considered sagittal plane motion without rotation or lateral bending contributions. A possible explanatory mechanism for this unexpected reduction in lumbo-sacral flexion with the tallest footrest may be the lumbar rotation and lateral bending coupling hypothesis (Panjabi, 1988). This hypothesis states that lumbar rotation is always associated with lateral bending motion. In vivo and in vitro analyses have shown that an axial torque applied to the neutral spine is associated with vertebral rotation and lateral flexion (Panjabi 1988; Shin et al., 2013). Studies interested in quantifying the degree of coupling and identifying segmental contributions where this motion occurs have produced conflicting results (Shin et al., 2013). However, this phenomenon is well accepted within the literature and may provide a compensatory mechanism by which our body can achieve more perceived global flexion through rotation and lateral bending.

A similar concept called the pelvic step may help to explain the same pattern that was observed in the lumbar to thigh angle on the elevated limb side. As individuals step forward, their pelvis rotates and positions the hip in a forward facing manner. As such, hip abduction and

external rotation can be utilized to increase perceived global hip flexion. The pelvic step has been shown to account for small increases in stride length during walking and running and it may be at play while standing with one foot elevated on a footrest (Liang et al., 2014; Sessoms, 2008; Whitcombe et al., 2017). Both of these mechanisms may contribute towards increased global flexion that cannot be captured through sagittal plane motion analysis alone.

Over time, all participants demonstrated an increase in lumbo-sacral flexion angle by 1-$2^0$ regardless of the condition. This may represent the early stages of viscoelastic creep. Creep refers to the progressive deformation of viscoelastic tissues following constant improper tissue loading such as when the lumbar spine is loaded in full flexion for a prolonged period of time (McGill & Brown, 1992). Twenty minutes of full lumbar flexion can result in $5.5^0$ increase of flexion due to viscoelastic hysteresis (McGill & Brown, 1992). Although the current study did not utilize full flexion, it may be considered a contributing factor after prolonged submaximal lumbar flexion.

When we compare the lumbar-to-thigh angles on both limbs, we see that flexion in the elevated limb was consistently offset by extension in the stance limb. The magnitude of angle change was extremely similar but opposite in direction (Figure 3.8). The net angle between both hips remained similar throughout all conditions and barely deviated from the baseline (net angle = $0.48^0$, SD = $0.14^0$). This suggests that individuals adopt a common net hip angle when using a footrest that is very similar to our anatomical norm. This concept may be important for maintaining optimal equilibrium during asymmetrical tasks. It is unclear if this trend is specific to standing with a footrest or if is consistent across all asymmetrical lower limb tasks.

Based upon the results of this study, we cannot conclude that one footrest height is preferable for all workers. The appropriate footrest height is likely task dependent and influenced by a number of factors including personal preference, environmental variations and anthropometric measures. In agreement with current recommendations, our findings suggest that a 10-20cm footrest should be used in the workplace. Additionally, when participants were asked to indicated their preferred footrest, they indicated the low and medium footrests were the most comfortable more often than the tallest footrest (i.e., 43.75%, 37.5% and 18.75% respectively).

The results of this study should be implemented with caution. The study focused on a non-clinical population but did not differentiate between PD's and NPD's. We know that PD's and

NPD's have been shown to respond differently when using a footrest and the benefits may not be experienced by all seemingly healthy individuals (Fewster et al., 2017). PD's may experience benefits but NPD's may actually be negatively affected when a footrest is used (Fewster et al., 2017). Future research should continue to examine these populations separately. Future research should also utilize a larger sample size with a more robust kinematic analysis of the lower limb. Kinematic analyses should utilize a flexion/extension – lateral bending rotation sequence to account for multi-axis motion (Fewster et al., 2017). Future studies should also examine how switching between feet during prolonged standing may affect the results.

In conclusion, this study aimed to identify if the presence of a footrest significantly affects joint position and muscle activation patterns and to examine if one specific footrest height could be recommended to all workers. The results of this study suggest that the use of a footrest does indeed affect muscle activation in the elevated limb and joint angles through the low back and both hip joint regions. The changes associated with using a footrest may provide workers with a tertiary option along with sitting and standing to maximize the benefits of a dynamic postural approach in the workplace. Occupational health and safety organizations should continue to recommend the use of a footrest during prolonged standing that measures between 10-20 cm. This acceptable variability allows individuals to account for personal preference and anthropometric differences that exist amongst all workers. Giving workers the ability to choose their own footrest height within this range may provide them with autonomy and self-confidence that improves overall compliance.

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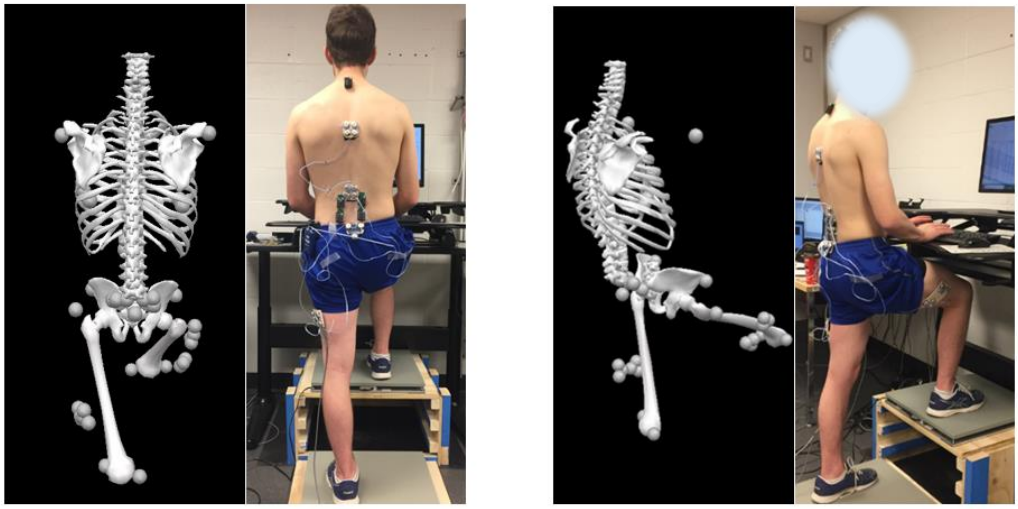

***Figure 3.1. Posterior view (left) and 45º posterior lateral view (right) of the experimental set up while using the highest footrest. Note the location of EMG electrodes and motion capture markers.***

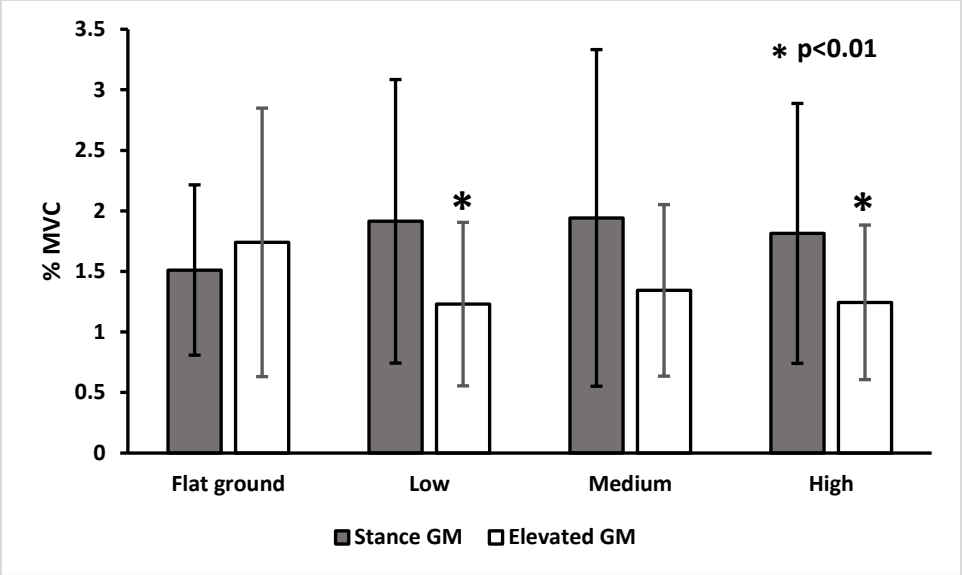


***Figure 3.2: %MVC of the GM muscles bilaterally across conditions. Note the main effect of footrest condition on the elevated limb GM. The asterisk demonstrates significantly lower muscle activity when compared to flat ground trial.***

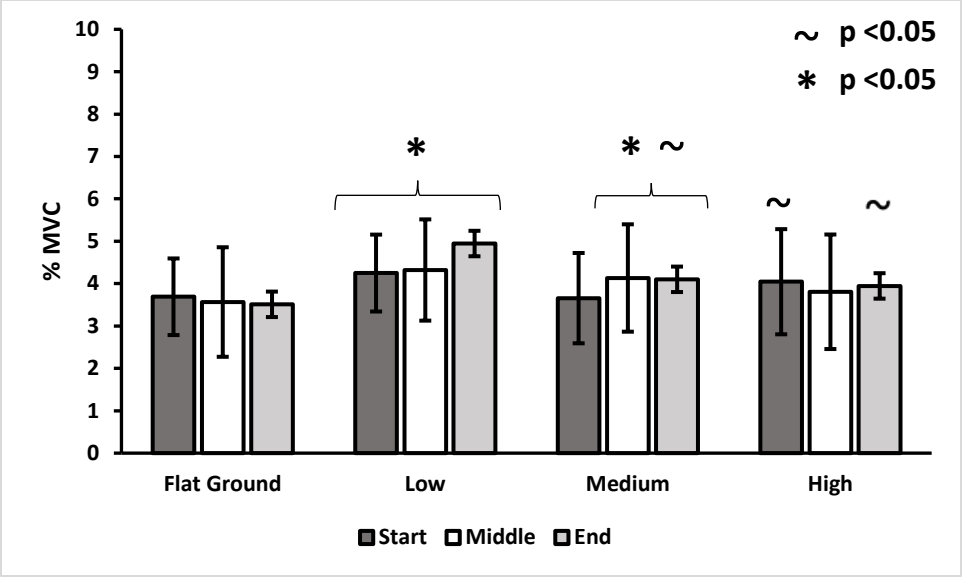


***Figure 3.3: Interaction effect between footrest height and time for the elevated limb ES. The asterisk demonstrates significantly higher activity compared to flat ground stance at specific time periods. The tilde (~) identifies significantly lower activity than the low footrest at specific time periods.***

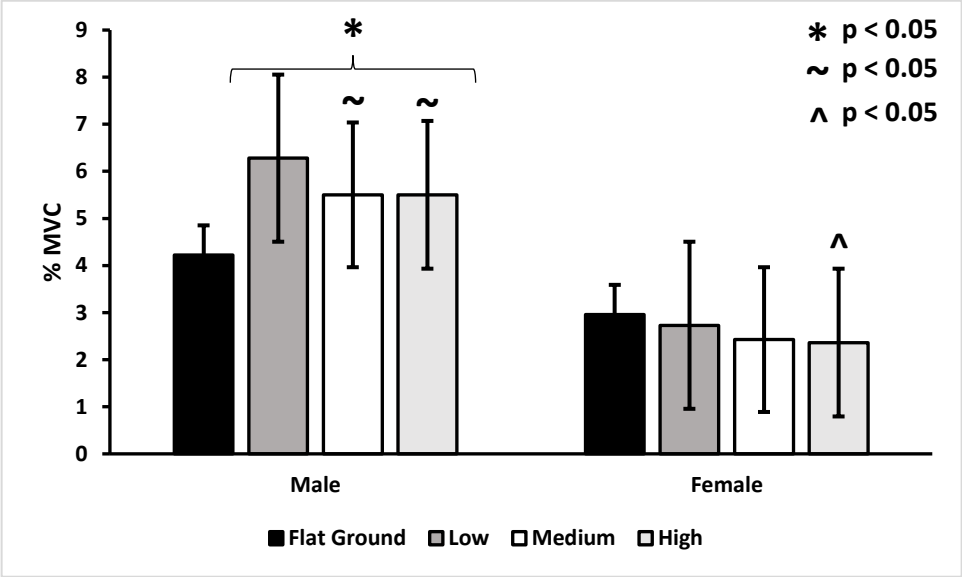


***Figure 3.4: Interaction effect of footrest condition and sex for the elevated ES. The asterisk demonstrates significant differences from flat ground. The tilde demonstrates significantly lower activity than the low footrest in males. The up arrow ("^") indicates significant difference compared to flat ground stance in females.***

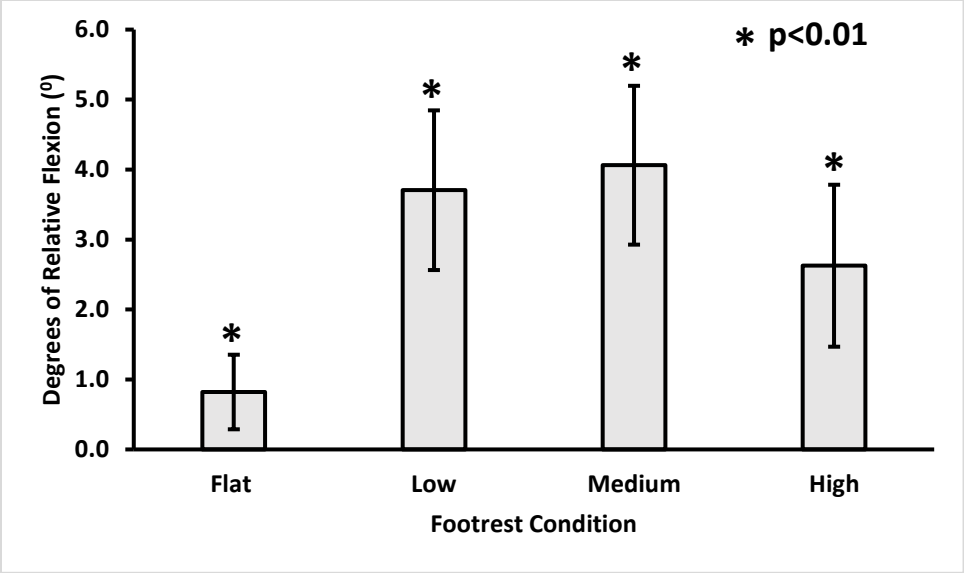


***Figure 3.5: Main effect of footrest condition on thoraco-sacral angle. The asterisk demonstrates significant pairwise comparisons (p<0.01).***

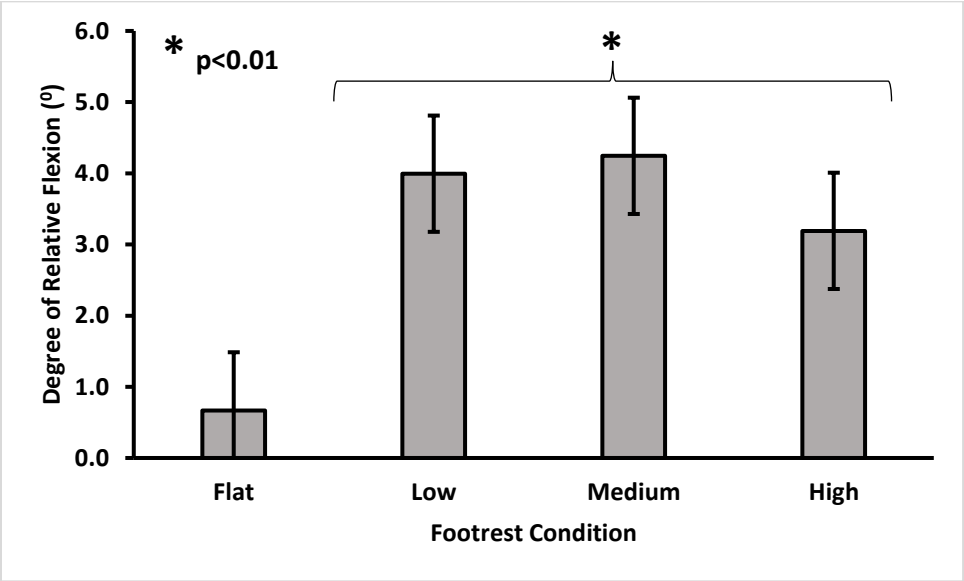


***Figure 3.6: Main effect of height on lumbo-sacral angle. The asterisk demonstrates significant differences compared to flat ground stance (p < 0.01).***

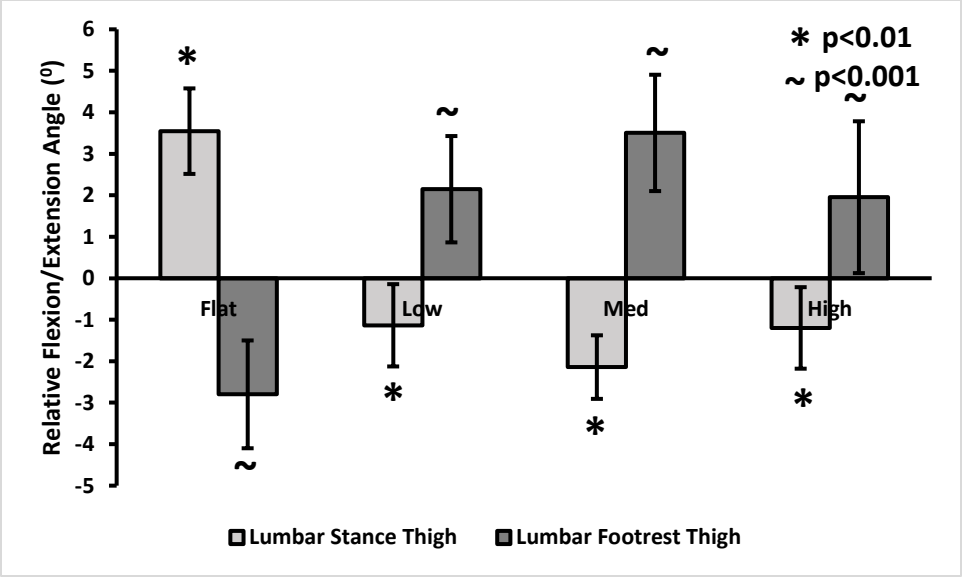


***Figure 3.7: Main effect of footrest condition on lumbar-to-thigh angles. The asterisk (p < 0.01) and tilde (p < 0.001) demonstrate significant comparisons within the lumbar stance thigh and lumbar footrest thigh angles respectively.***

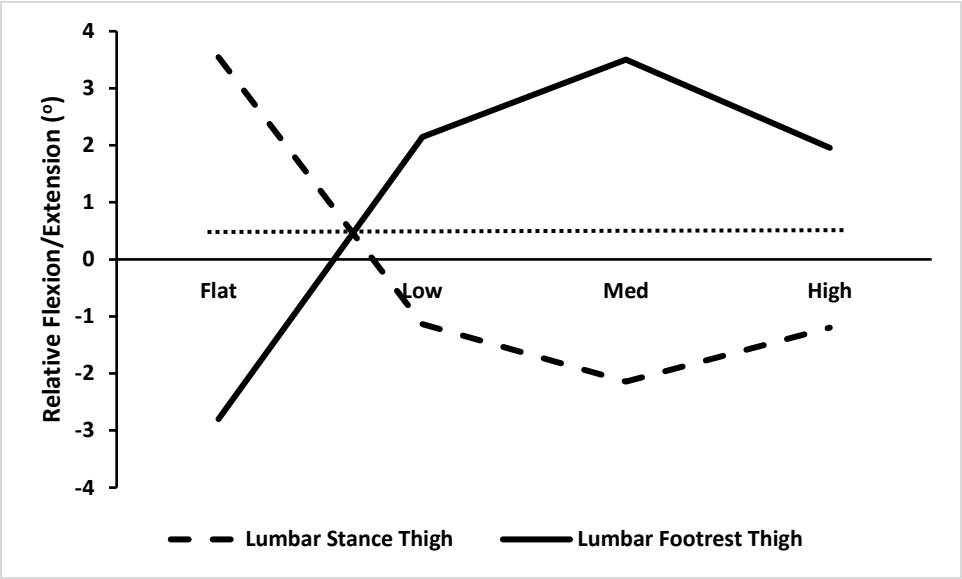


***Figure 3.8: Relationship between left and right thigh joint angle across footrest condition with the dotted line representing the overall average net joint angle between stance and elevated limbs.***

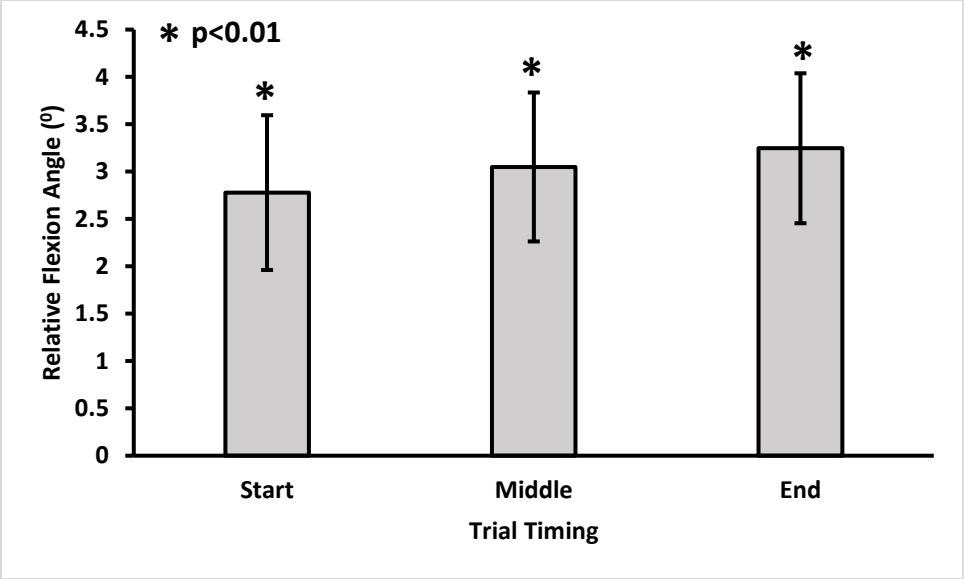


***Figure 3.9: Main effect of time on lumbo-sacral angle. The asterisk demonstrates a significant increase in flexion over time regardless of condition or sex (p < 0.01).***

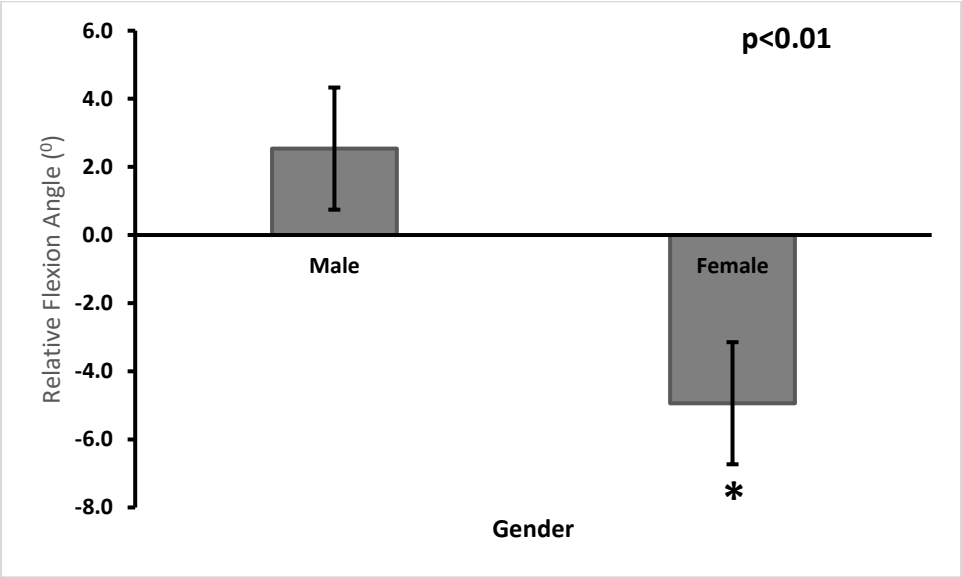


***Figure 3.10: Significant main effect of sex on lumbar to footrest thigh angle. Males stood in an overall flexion position whereas females stood with extension.***

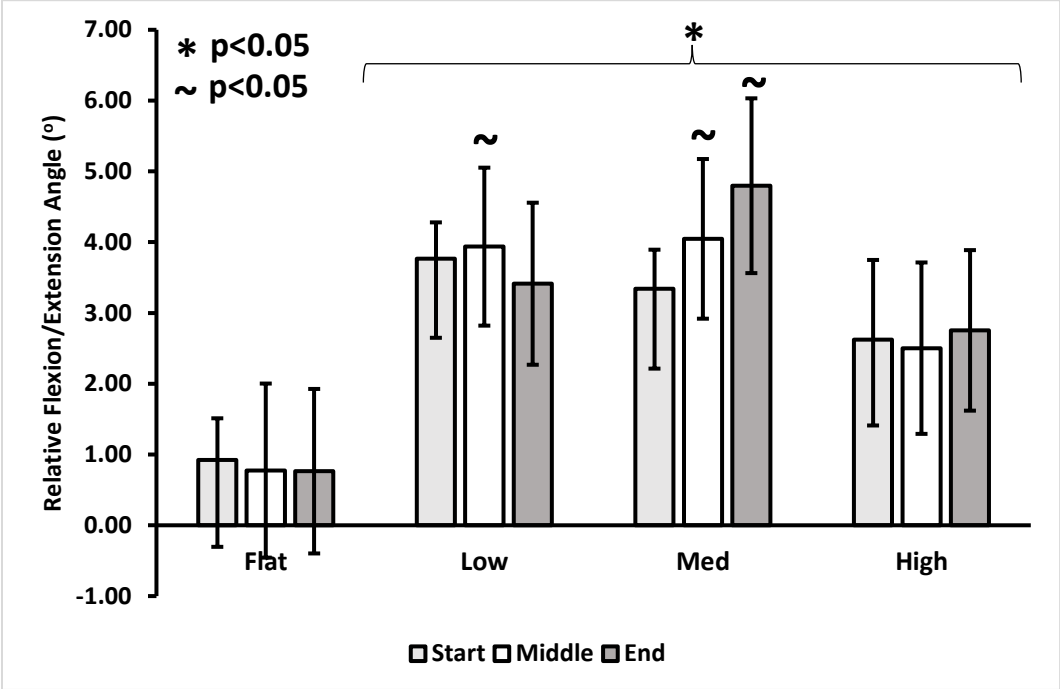


***Figure 3.11: Interaction effect of footrest condition and time for the thoraco-sacral angle. The asterisk indicates significant differences when compared to flat ground trials and the tilde demonstrates significant differences when compared to the high footrest trial at the same time period***

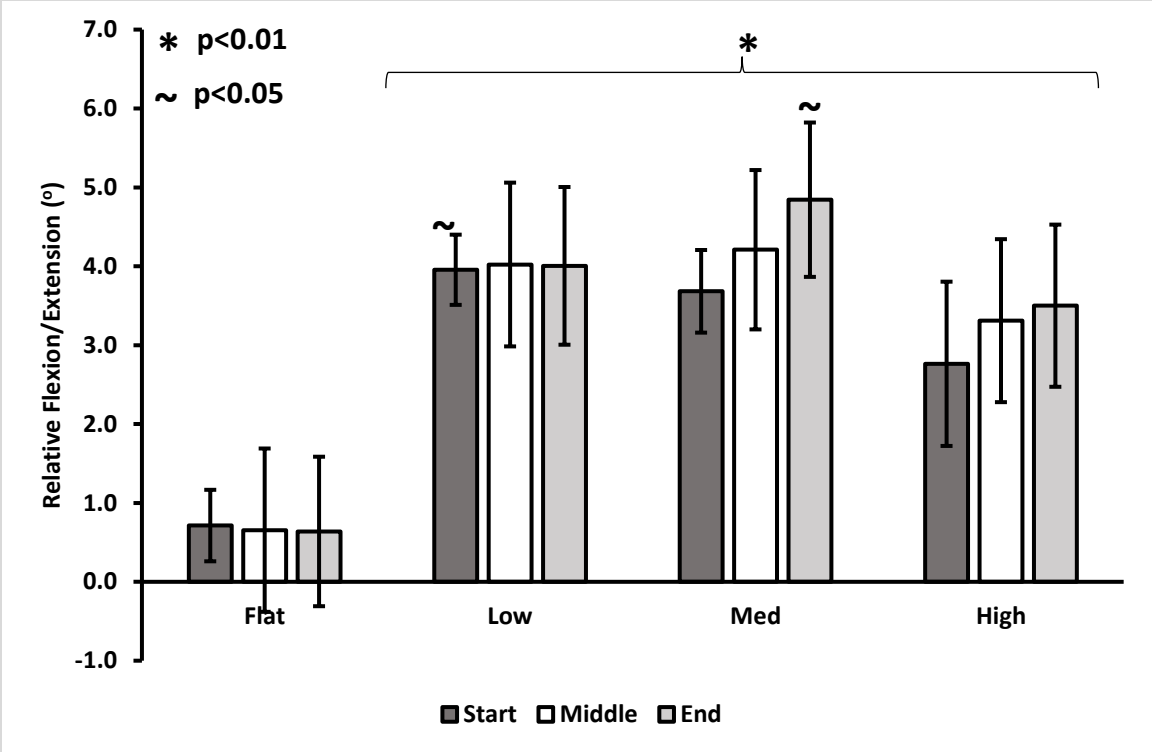


***Figure 3.12: Interaction effect of footrest condition and time for the lumbo-sacral angle. The asterisk demonstrates significantly more flexion in the lumbo-sacral region when compared to flat ground stance. The tilde demonstrates significantly higher lumbo-sacral flexion when compared to the highest footrest at the same time period.***

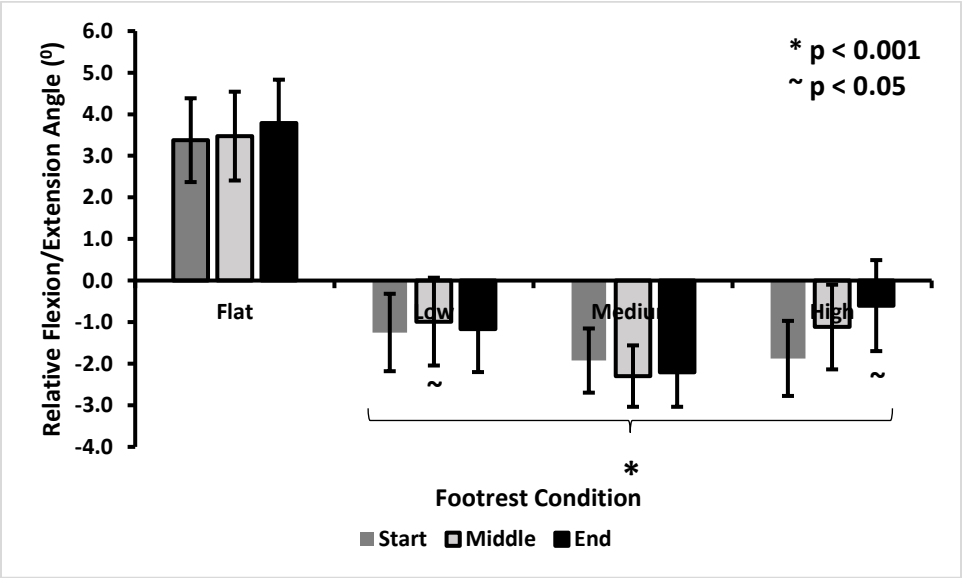


***Figure 3.13: Interaction effect of footrest condition and time for the lumbar to stance thigh angle. The asterisk demonstrates differences from flat ground. The tilde demonstrates significant differences from the medium footrest at the specified time period.***

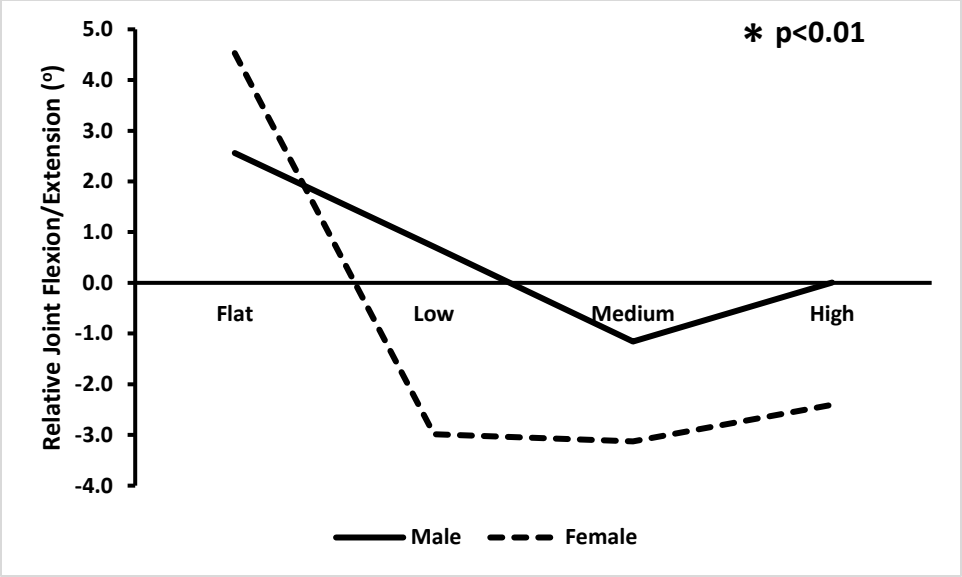


***Figure 3.14: Interaction effect between footrest condition and sex for lumbar to stance thigh angle. Positive angles are flexion, negative represent extension.***

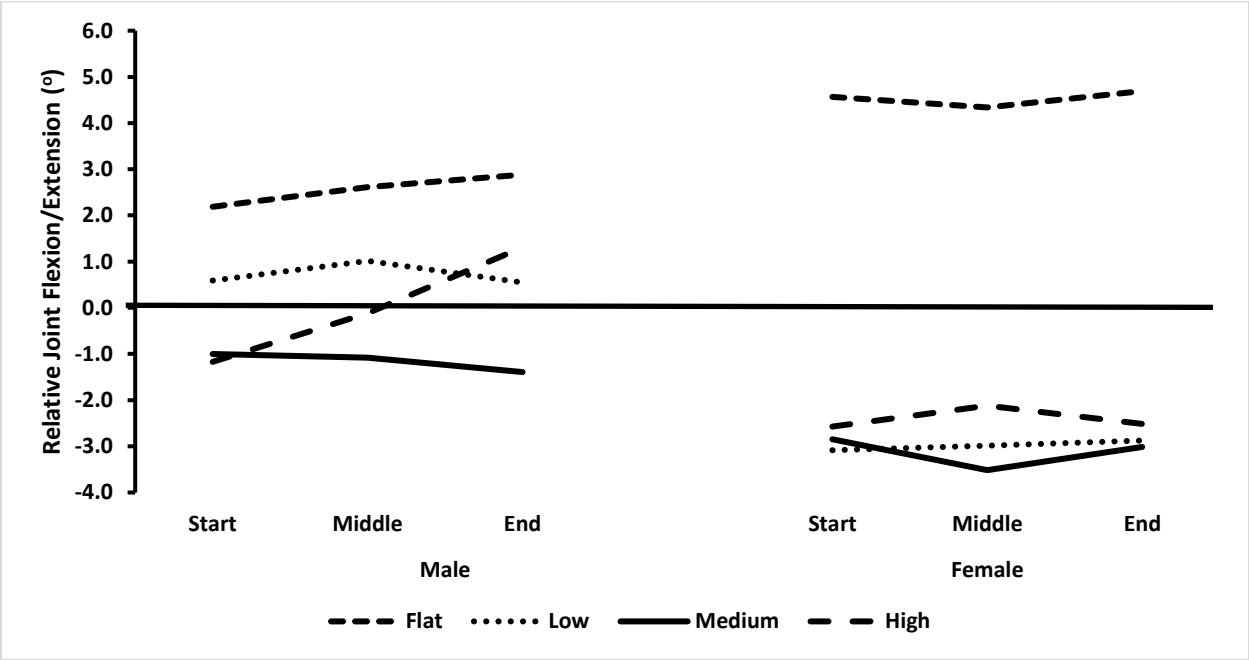


***Figure 3.15: Three-way interaction effect of footrest height*time*sex for lumbar to stance thigh angle. Positive angles represent flexion whereas negative angles represent extension.***

**Table 3.1: Demographic Information of Male and Female Participants including Means ($\bar{x}$) and Standard Deviations (SD) of Age, Height, Weight and Written Outcome Measure Scores.**

| | Male | | Female | | Combined | |
|---|---|---|---|---|---|---|
| | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD |
| Age: | 24.88 | 4.190 | 23.38 | 4.75 | 24.13 | 4.40 |
| Height: | 181.63 | 4.72 | 165.81 | 7.88 | 173.72 | 10.30 |
| Weight: | 81.16 | 12.88 | 60.12 | 10.49 | 173.72 | 15.72 |
| WHQ Score: | 28.29 | 34.94 | 56.88 | 14.02 | 43.53 | 28.98 |
| WFQ Score: | 5.57 | 9.50 | 11.13 | 8.20 | 8.53 | 8.98 |
| VAS: | 0.29 | 0.76 | 0.38 | 0.74 | 0.33 | 0.72 |
| OBDI Score: | 1.14 | 1.68 | 1.63 | 3.29 | 1.40 | 2.59 |

**Table 3.2: Mean ($\bar{x}$) and Standard Deviation (SD) of EMG Activity (%MVC) by Muscle Group, Limb, Footrest Condition, and Time. Muscles Include the Lumbar Erector Spinae (ES), Gluteus Medius (GM) and Tensor Fascia Lata (TFL) Bilaterally.**

| | | Stance ES | | Elevated ES | | Stance GM | | Elevated GM | | Stance TFL | | Elevated TFL | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD |
| Flat Ground | Start | 7.4 | 16.3 | 3.7 | 3.5 | 1.5 | 0.7 | 1.8 | 1.2 | 1.0 | 1.0 | 0.8 | 0.7 |
| | Middle | 7.7 | 16.9 | 3.6 | 3.6 | 1.5 | 0.7 | 1.8 | 1.2 | 0.9 | 0.9 | 0.8 | 0.9 |
| | End | 7.2 | 16.3 | 3.5 | 4.2 | 1.6 | 0.8 | 1.6 | 1.0 | 1.2 | 1.3 | 0.7 | 0.6 |
| | Full | 7.4 | 16.5 | 3.6 | 3.8 | 1.5 | 0.7 | 1.7 | 1.1 | 1.0 | 1.1 | 0.8 | 0.7 |
| Low Footrest | Start | 8.0 | 18.2 | 4.2 | 5.1 | 1.9 | 1.0 | 1.2 | 0.7 | 1.3 | 1.3 | 1.0 | 1.3 |
| | Middle | 8.0 | 19.4 | 4.3 | 4.8 | 1.9 | 1.2 | 1.2 | 0.7 | 1.5 | 1.8 | 1.0 | 1.1 |
| | End | 9.4 | 20.0 | 4.9 | 5.9 | 1.9 | 1.3 | 1.3 | 0.7 | 1.6 | 1.7 | 1.1 | 1.4 |
| | Full | 8.5 | 19.2 | 4.5 | 5.3 | 1.9 | 1.2 | 1.2 | 0.7 | 1.5 | 1.6 | 1.1 | 1.3 |
| Medium Footrest | Start | 3.9 | 4.7 | 3.7 | 4.2 | 1.9 | 1.4 | 1.3 | 0.7 | 1.2 | 1.4 | 1.1 | 1.2 |
| | Middle | 2.5 | 2.7 | 4.1 | 4.9 | 1.8 | 1.3 | 1.4 | 0.7 | 1.8 | 2.4 | 1.3 | 2.0 |
| | End | 2.7 | 2.8 | 4.1 | 5.3 | 2.1 | 1.5 | 1.4 | 0.7 | 1.4 | 1.5 | 1.0 | 1.0 |
| | Full | 3.1 | 3.4 | 4.0 | 4.8 | 1.9 | 1.4 | 1.3 | 0.7 | 1.5 | 1.8 | 1.1 | 1.4 |
| High Footrest | Start | 7.3 | 17.7 | 4.0 | 4.9 | 1.8 | 1.1 | 1.2 | 0.6 | 1.4 | 1.6 | 1.7 | 2.4 |
| | Middle | 7.5 | 19.4 | 3.8 | 5.1 | 1.7 | 1.0 | 1.2 | 0.6 | 1.4 | 1.5 | 0.9 | 0.7 |
| | End | 7.3 | 18.6 | 3.9 | 5.5 | 1.9 | 1.2 | 1.3 | 0.7 | 1.2 | 1.5 | 1.0 | 1.0 |
| | Full | 7.4 | 18.6 | 3.9 | 5.2 | 1.8 | 1.1 | 1.2 | 0.7 | 1.3 | 1.5 | 1.2 | 1.4 |

**Table 3.3: Interaction and Main Effects for EMG Activity for the Erector Spinae (ES), Gluteus Medius (GM) and Tensor Fascia Lata (TFL). Significant Tests are Denoted by an Asterisk.**

| | Stance ES | | Elevated ES | | Stance GM | | Elevated GM | | Stance TFL | | Elevated TFL | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | F | Sig | F | Sig | F | Sig | F | Sig | F | Sig | F | Sig |
| Footrest Condition | 1.268 | 0.282 | 2.224 | 0.141 | 2.608 | 0.088 | 8.257 | **0.007*** | 2.059 | 0.153 | 1.375 | 0.268 |
| Time | 0.330 | 0.641 | 0.852 | 0.416 | 1.318 | 0.280 | 1.069 | 0.346 | 1.542 | 0.236 | 3.030 | 0.097 |
| Sex | 0.701 | 0.416 | 1.439 | 0.250 | 0.166 | 0.690 | 2.811 | 0.116 | 2.752 | 0.119 | 0.029 | 0.868 |
| Footrest Condition*Time | 1.243 | 0.305 | 3.145 | **0.023*** | 0.341 | 0.840 | 1.409 | 0.255 | 1.121 | 0.334 | 1.292 | 0.282 |
| Footrest Condition*Sex | 1.175 | 0.300 | 4.011 | **0.042*** | 0.745 | 0.490 | 1.199 | 0.303 | 0.518 | 0.580 | 0.980 | 0.366 |
| Time*Sex | 0.273 | 0.680 | 1.223 | 0.304 | 2.728 | 0.102 | 0.252 | 0.731 | 0.342 | 0.598 | 0.180 | 0.712 |
| Condition*Time*sex | 1.419 | 0.258 | 1.844 | 0.136 | 0.833 | 0.505 | 0.563 | 0.636 | 0.924 | 0.396 | 0.899 | 0.379 |

**Table 3.4: Mean ($\bar{x}$) and Standard Deviation (SD) of Relative Flexion/Extension by Joint Angle, Footrest Condition and Time. All Values are Measured in Degrees where Positive Values Demonstrate Relative Flexion and Negative Angles Represent Relative Extension from Baseline.**

| | | Thoraco-sacral | | Thoraco-sacral | | Lumbo-sacral | | Lumbar Stance Thigh | | Lumbar Footrest Thigh | |
|---|---|---|---|---|---|---|---|---|---|---|---|
| | | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD |
| Flat Ground | Start | 0.53 | -1.66 | 0.93 | -2.27 | 0.71 | -1.75 | 3.38 | -4.08 | -2.91 | -5.48 |
| | Middle | 0.27 | -1.79 | 0.77 | -1.99 | 0.65 | -1.72 | 3.47 | -4.22 | -3.01 | -5.45 |
| | End | 0.34 | -1.75 | 0.76 | -2.13 | 0.64 | -2.03 | 3.79 | -4.14 | -2.48 | -5.33 |
| | Full | 0.38 | -1.73 | 0.82 | -2.13 | 0.67 | -1.83 | 3.55 | -4.15 | -2.80 | -5.42 |
| Low Footrest | Start | 0.51 | -3.18 | 3.76 | -4.38 | 3.96 | -4.12 | -1.25 | -4.07 | 2.38 | -5.70 |
| | Middle | 0.31 | -3.35 | 3.94 | -4.83 | 4.02 | -4.17 | -0.99 | -4.58 | 2.12 | -6.10 |
| | End | 0.02 | -3.13 | 3.41 | -4.40 | 4.01 | -4.21 | -1.17 | -4.37 | 1.94 | -5.93 |
| | Full | 0.28 | -3.22 | 3.70 | -4.54 | 3.99 | -4.17 | -1.13 | -4.34 | 2.15 | -5.91 |
| Medium Footrest | Start | 0.60 | -3.28 | 3.34 | -4.38 | 3.68 | -4.10 | -1.92 | -3.13 | 3.37 | -7.29 |
| | Middle | 0.96 | -3.56 | 4.05 | -4.69 | 4.21 | -4.12 | -2.30 | -3.12 | 3.53 | -7.05 |
| | End | 1.10 | -3.57 | 4.80 | -4.51 | 4.84 | -3.77 | -2.20 | -3.32 | 3.61 | -7.38 |
| | Full | 0.89 | -3.47 | 4.06 | -4.53 | 4.25 | -3.99 | -2.14 | -3.19 | 3.50 | -7.24 |
| High Footrest | Start | 1.37 | -2.93 | 2.62 | -4.43 | 2.76 | -3.92 | -1.87 | -3.56 | 1.52 | -9.07 |
| | Middle | 0.52 | -3.39 | 2.50 | -4.78 | 3.31 | -3.85 | -1.12 | -4.07 | 1.99 | -9.24 |
| | End | 0.63 | -2.96 | 2.75 | -4.41 | 3.50 | -4.06 | -0.60 | -4.67 | 2.35 | -8.98 |
| | Full | 0.84 | -3.09 | 2.63 | -4.54 | 3.19 | -3.94 | -1.20 | -4.10 | 1.95 | -9.10 |

**Table 3.5: Interaction and Main Effect Results for Each Joint Angle. Significant Values are Denoted by an Asterisk.**

| | Thoraco-sacral | | Thoraco-sacral | | Lumbo-sacral | | Lumbar Stance Thigh | | Lumbar Footrest Thigh | |
|---|---|---|---|---|---|---|---|---|---|---|
| | F | p-value | F | p-value | F | p-value | F | p-value | F | p-value |
| Footrest Condition | 0.822 | 0.432 | 7.175 | **0.004*** | 11.613 | **0.001*** | 16.502 | **0.000*** | 10.268 | **0.001*** |
| Time | 0.768 | 0.446 | 0.898 | 0.409 | 7.762 | **0.004*** | 1.899 | 0.184 | 1.309 | 0.286 |
| Sex | 0.025 | 0.877 | 0.073 | 0.791 | 0.828 | 0.378 | 0.968 | 0.342 | 8.694 | **0.011*** |
| Footrest Condition *Time | 2.043 | 0.105 | 3.169 | **0.028*** | 2.827 | **0.045*** | 3.027 | **0.040*** | 1.151 | 0.338 |
| Footrest Condition *Sex | 0.987 | 0.374 | 0.404 | **0.656** | 0.919 | 0.387 | 3.761 | **0.044*** | 3.359 | 0.056 |
| Time*Sex | 0.307 | 0.686 | 0.162 | 0.826 | 0.193 | 0.794 | 1.612 | 0.225 | 0.152 | 0.854 |
| Footrest Condition *Time*sex | 1.148 | 0.343 | 1.533 | 0.215 | 0.757 | 0.533 | 3.089 | **0.037*** | 0.956 | 0.414 |

# CHAPTER 4: USE OF A FOOTRET WHILE WORKING AT A STANDING WORKSTATION AND ITS EFFECT ON WEIGHT DISTRIBUTION, CENTRE OF PRESSURE (COP) AND DISCOMFORT MEASURES

**Andrew C. Cregg, Ryan C.A. Foley, Nicolas J. La Delfa & Lori A. Livingston**

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

Corresponding Author

Lori A. Livingston, PhD

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

2000 Simcoe St. N.

Oshawa, ON, Canada

L1H 7K4

Telephone: (905)721-8668

Email: Lori.Livingston@uoit.ca

**Abstract**

Footrests are a commonly recommended standing aid to help reduce the risk of developing low back pain (LBP). Few studies have examined how footrest usage impacts postural stability and discomfort. Even fewer studies have examined how the height of a footrest affects these factors. Given the wide variety of footrests available today, this is an important factor to examine so that workers can select a footrest that optimizes their balance and minimizes discomfort. The purpose of this study was to examine how the presence of a footrest affects balance and discomfort and to determine if the relationship is dependent upon footrest height. Participants discomfort was measured using a Visual Analogue Scale (VAS) and Rating of Perceived Discomfort (RPD). Balance was examined using a dual force platform design.

Fifteen healthy participants were exposed to four trials lasting 15 minutes each while completing a standardized computer task at a standing desk. The trials included standing on flat ground, standing with a low (10 cm), medium (20 cm) and high (30 cm) footrest. The results from the study suggest that footrest usage shifted percentage weight distribution from a symmetrical position to an asymmetrical positon that was not dependent upon footrest height. Regardless of footrest height, approximately 85% of the weight was distributed to the stance limb and the remaining 15% was placed on the elevated limb. Sway magnitude was higher in the elevated limb in both the anterior/posterior (A/P) and medial/lateral (M/L) directions and was dependent upon footrest height. Discomfort increased as a function of time regardless of condition.

Therefore, footrest usage altered percentage weight distribution by shifting the majority of weight to the stance limb. Sway increased in the elevated limb during footrest trials and was dependent upon footrest height such that higher footrests induced more sway. From a discomfort perspective, standing for 15 minutes was associated with increases in discomfort regardless of footrest height. The presence of a footrest did not significantly increase discomfort and should be considered as an acceptable intervention for healthy individuals.

**Key Words:**

Footrest, low back pain (LBP), ergonomics, standing desk, sit-stand desk, force, kinetics, discomfort, Visual Analogue Scale, footrest

**Statement of Authorship:**

Andrew C. Cregg (Masters of Health Science Candidate) was the primary investigator for this investigation and handled all aspects of the research agenda.

## Introduction

The self-reported three month prevalence of low back pain (LBP) amongst workers is 25.7% in the United States, 28.7% in Canada and 18% in the United Kingdom (Yang, Haldeman, Lu & Baker, 2017). This corresponds to an overall average of approximately one quarter ($\bar{x}$ = 24.11%) of workers being exposed to LBP over a 3 month period in these countries. From a global perspective, LBP has been identified as the leading cause of disease and years lived with disability worldwide, contributing significantly toward overall global disability and mortality rates (Driscoll et al., 2014; Vos et al., 2015). Occupationally induced LBP has traditionally been associated with labour intensive occupations. However, a growing body of evidence suggests that workers who maintain prolonged static postures such as sitting and standing are also at an increased risk of developing LBP (Callaghan & McGill, 2001). Under both sitting and standing conditions, prolonged static loads of two hours or more could result in injury due to fatigue (Callaghan & McGill, 2001). Prolonged standing results in fatigue to the active stabilizing system whereas sitting places undue stress to the passive stabilizing system (Callaghan & McGill, 2001). Therefore, neither standing nor sitting are ideal postures that should be held for prolonged periods of time. Most experts agree that workers should begin by replacing periods of sitting with intermittent standing or other light activities. This will reduce overall sedentary time and allow workers to take advantage of variable postures throughout their day (Buckley et al., 2015; Coenen et al., 2017).

A dynamic postural approach takes advantage of both sitting and standing throughout the day to promote postural, muscular and balance variability while improving fluid tissue dynamics. Alternating between standing and sitting allows workers to cyclically load the active and passive musculoskeletal systems respectively while providing intermittent rest breaks to the inactive system (Callaghan & McGill, 2001). This method allows individuals to disperse the load across the spectrum of tissues rather than applying prolonged stress to any one tissue type. It also allows postural variability that takes advantage of relative changes in lumbar flexion and extension that may lead to a reduction in LBP (Gallagher et al, 2011).

In order to identify strategies to help reduce the risk of LBP associated with sitting and standing, it is important to examine our current understanding of these postures. Prolonged sitting is a sedentary behavior that leads to low caloric expenditure and increased risk of preventable chronic health conditions (Frymoyer, 1980; Reiff et al., 2012). Oxygen consumption ($VO_2$), carbon dioxide production ($VCO_2$), and caloric expenditure are higher after a 45 minute prolonged standing period when compared to a prolonged seated period (Reiff et al., 2012). For this reason, occupational health and safety authorities have targeted occupational sitting time as a way of changing caloric expenditure in the workplace and to reduce overall sedentary time.

Prolonged standing also places workers at an increased risk of developing certain preventable chronic health conditions, including musculoskeletal disorders such as LBP and lower extremity symptoms (Coenen, 2017; Lee et al., 2018). Approximately 40-64% of otherwise healthy individuals will develop LBP following periods of prolonged standing (Nelson-Wong & Callaghan, 2010). A dichotomous classification system has been developed that differentiates between pain developers (PD's) and non-pain developers (NPD's) depending upon their likelihood of developing LBP following prolonged standing despite having no previous history of LBP (Nelson-Wong & Callaghan, 2010). Workers who have no prior history of LBP may still be at an increased risk of developing it after prolonged standing. PD's are individuals with no previous history of LBP but who develop it after approximately 42 minutes of prolonged standing (Nelson-Wong, Gregory, Winter & Callaghan, 2008). As such, standing should be recommended with caution in healthy individuals and workers should be encouraged to adopt a dynamic postural approach throughout the day as a preventative measure.

Sit-stand desks are often recommended as an intervention to promote a dynamic postural approach in the workplace. Sit-stand desks allow workers to assume various positions throughout the work day and provide them with the autonomy to choose their posture. However, defining exposure limits for prolonged standing has become a topic of debate in the literature. A recent systematic review examined pooled dose-response associations between prolonged standing and LBP (Coenen et al., 2017). They found that

clinically relevant levels of LBP begin after 71 minutes of prolonged standing in healthy individuals and after only 42 minutes in PD's. Overall, they recommend that 40 minutes should be set as the standardized upper limit of prolonged standing to protect all workers (Coenen et al., 2017).

Researchers also continue to explore the utility of other commonly used interventions to further improve comfort and reduce the risk of injury in the workplace. A primary goal of these studies is to identify simple, inexpensive, and effective interventions that can be used in conjunction with sit-stand desks. Research efforts have focused on various standing aids including anti-fatigue mats (Wiggermann & Keyserling, 2013), sloped surfaces (Fewster et al., 2017; Nelson-Wong & Callaghan, 2010), staggering stance (Fewster et al., 2017) and elevating one foot on a footrest (Fewster et al., 2017). Fewster and colleagues (2017) compared the effects of four commonly used standing postures on lumbar kinematics and muscle activation patterns during short term standing periods. They examined standing on flat ground, standing on a sloped surface, standing with a staggered stance, and using a single legged footrest. The footrest was the only effective intervention observed to alter lumbar spine posture and gluteus medius co-activation patterns in comparison to flat ground stance. In fact, multiple studies have reported that the use of a footrest can significantly alter lumbar posture, muscle activation patterns, and postural stability (Dolan, 1988; Fewster et al., 2017; Lee et al., 2018; Mohan et al.,2014), while improving comfort levels (Rys and Konz, 1994).

The benefits associated with using a footrest may be in part due to the asymmetrical nature of the position. While standing quietly on a flat surface, the body assumes a symmetrical posture with both feet planted firmly on the ground. Symmetrical stance is defined as loading 50% +/- 15% (mean, +/- SD) of an individual's body weight to both limbs (Mohan et al., 2014; Wang et al., 2012). When using a standing aid such as a footrest, the body is placed in an asymmetrical posture with unique loading profiles. Asymmetrical stance is defined as supporting two-thirds (i.e., greater than 65%) of an individual's body weight on one leg with the remaining weight on the other leg (Anker et al., 2008; Gallagher, Nelson-Wong & Callaghan, 2011). Examples of common

asymmetrical positions include tandem stance and staggering one's feet on a 45 degree angle. In these positions, 65-75% of an individual's body weight is loaded to the back limb suggesting a supportive role of the hind limb (Genthon & Rougier, 2005; Wang et al., 2012). When a 15 cm footrest is used, participants place approximately 80% of their weight through the hind limb and 20% through the elevated limb (Mohan et al., 2014). This 80%/20% split was consistent when the footrest was placed in front or beside the individual (Mohan et al., 2014). Asymmetrical stance has also been associated with increased body sway and changes in centre of pressure (COP) location over time (Anker et al., 2008; Wang & Newell, 2012). COP mean distance, Root Mean Square (RMS), COP range, and COP velocity increase significantly in the anterior/posterior (A/P) and medial/lateral (M/L) direction when individuals used a footrest (Mohan et al., 2014). Again, these differences were observed when the footrest was placed in front and beside the participants, and the effects were magnified when visual feedback was removed. However, the majority of these observations were generated during short, 30 second trials and may differ from what would be observed during periods of prolonged standing.

To our knowledge, only one study to date has examined how differences in footrest height affect body mechanics through observation of lumbo-pelvic joint angles, external lumbar moments, erector spinae fatigue, and subjective pain levels (Son et al., 2017). This investigation examined the aforementioned dependent variables during three separate footrest conditions (i.e., footrest height was adjusted to 5%, 10% and 15% of an individual's body height) compared to flat ground stance. Results from the study indicate that all footrest conditions resulted in less pain than flat ground stance, but the 10% footrest also minimized the external moment in the lumbar region and produced the smallest change in mean power frequency ratio. This led the authors to conclude that a footrest adjusted to 10% body height minimizes fatigue, discomfort, and external load to the lumbar spine (Son et al., 2017). However, this study was conducted on a clinical population with non-specific LBP and the results therefore are not generalizable to a healthy population. Additionally, they utilized normalized recommendations based upon each individual worker's height. We believe that providing normalized values may lead to

poor compliance and low ecological validity. Workers and employers are more likely to implement footrests when given an absolute value rather than taking measurements to adjust the footrest out of convenience and standardization.

It is important to consider subjective comfort as an indicator of compliance and adherence when implementing new workplace interventions. Participants report that they are more comfortable when using a 10 cm platform during a four hour standing period when compared to standing on flat ground (Rys & Konz, 1989). Studies commonly use the Visual Analogue Scale (VAS) and Rating of Perceived Discomfort (RPD) scales to quantify subjective discomfort. The VAS utilizes a 100 mm scale that allows individuals to mark their level of discomfort. The minimum clinically significant difference for the VAS has been reported as 9 mm (Kelly, 1998).

Despite the relative lack of evidence to support their use, footrests have become the most commonly recommended improvised standing aid by occupational health and safety organizations (Fewster et al., 2017). However, very little is known about the appropriate specifications that would maximize the benefits associated with footrest usage. Current recommendations suggest that workers use a 10-20 cm footrest (Ebben, 2003). However, the current recommendations are vague and do not provide adequate guidelines including appropriate dimensions or placement of the footrest. The height of a footrest is one such factor that may significantly impact balance and discomfort during prolonged periods of standing. Utilization of a footrest places the body in a uniquely asymmetrical posture that is not currently well understood. Percentage of weight distribution and centre of pressure (COP) are two important kinetic variables that can help us understand how balance and postural stability are impacted during asymmetrical stance positions.

Therefore, the purpose of this study was to investigate how the height of a footrest affects discomfort, balance, weight distribution and postural sway in comparison to flat ground stance while individuals work at a standing workstation. It was hypothesized *a priori* that the use of a footrest would increase the amount of weight

distributed to the support limb and decrease the weight distributed to the elevated limb. It was further hypothesized that the degree of change would be dependent upon the height of the footrest such that more weight would be placed on the stance limb as the height of the footrest increased. From a postural stability perspective, it was also hypothesized that sway would increase in the A/P and M/L direction under both limbs as the height of the footrest increased. From a discomfort perspective, it was expected that the highest and lowest footrest would result in the most and least discomfort, respectively. Importantly, the overarching aim of this study was to inform occupational health and safety recommendations concerning the use of a footrest during prolonged standing tasks.

## Methods

*Participants:*

Fifteen healthy participants (i.e., 7 male and 8 female, age = 24.3 years +/- 4.5, height = 173.1 cm +/- 10.3 cm, mass = 70.8 kg, +/- 16.3 kg) were recruited from the University of Ontario Institute of Technology (UOIT) population based upon a power calculation ($\alpha = 0.05$, $\beta = 0.80$) to reduce Type II error. Participants were excluded if they reported a history of LBP that required medical intervention or time off work greater than three days in the last 12 months, employment in a job that required prolonged static standing over the past 12 months, had previous lumbar or hip surgery, an inability to stand for two hours or a clinically significant score on the Oswestry Low Back Disability Index (OBDI) (Fewster et al., 2017; Gallagher, 2014, Nelson-Wong & Callaghan, 2010). Participants completed the Waterloo Handedness and Footedness questionnaires, the OBDI, and they provided demographic information prior to participation (Appendix A-F). Prior to participating in the study, individuals deemed eligible for inclusion within the study provided both written and verbal informed consent. The study was approved by the UOIT research Ethics Board (REB# 14477).

*Instrumentation:*

Two portable force platforms (AccuGait 30lb Max Range, AMTI) were used to collect ground reaction force data associated with each limb using AMTI NetForce Software™ (AMTI, Watertown, USA). Force data were collected at a frequency of 1000Hz and both platforms were time synchronized. During flat ground trials, both platforms were positioned side-by-side with one limb placed on either platform. During footrest trials, one platform was positioned on top of the custom made footrest while the other remained on the floor. Each force platform was fastened to a wooden stage that allowed for quick and stable switching between footrest heights (Figure 4.1).

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Insert Figure 4.1 Here

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The standing desk was modified to allow 5-6 cm of clearance between the participant's wrists and the table when their elbows were placed at 90 degrees flexion (Kroemer & Grandjean, 1997). Participants discomfort was objectified using the VAS and RPD scales (Appendix F). The minimum clinically significant difference for VAS score was set at 10 mm to align with previous similar research (Lee et al., 2018; Nelson-Wong & Callaghan, 2010). Participants were asked to complete the discomfort questionnaire at minute 1, 5, 10 and 15 during each level ground and footrest condition trial (i.e., 10, 20, 30 cm footrest height) resulting in a total of 16 scores for each participant. Scores recorded within the first minute of each trial acted as the baseline measure. The other three scores were labelled as the "start", "middle" and "end" respectively.

*Experimental Protocol:*

All data were collected in conjunction with a concurrent study that analyzed muscle activation and joint position during the same task (see Chapter 3). Following the intake process and administration of consent, normalization procedures were conducted to measure participants' resting body weight, mass, and discomfort. Individual body mass was measured by standing on a single force platform for 5 seconds in anatomical position.

Each participant's height was then measured. The force platforms were time synchronized with the electromyographic (EMG) and kinematic measurements associated with the secondary study (see Chapter 3) and were triggered by a single keystroke.

The study involved four trials that lasted 15 minutes each and were separated by 3-5 minute rest breaks. Trials included flat ground stance, standing with a low footrest (10cm), a medium footrest (20cm), and a high footrest (30cm). For each participant, data collection lasted for approximately 2-2.5 hours. Flat ground stance was always the first trial followed by the other three trials in a randomized order. Participants raised their dominant foot on the footrest based upon the results of their Waterloo Handedness and Footedness Questionnaires. All participants except for one was right side dominant. Data for this participant were converted to align with the "stance limb" and "elevated limb."

Prior to each trial, participants were instructed to stand in their usual manner for the entire period of time. When a footrest was provided, participants were asked to maintain contact between their feet and the force platforms at all times. They were instructed to avoid leaning on the table and to not lift their foot off of the platform surfaces (Gallagher, 2014). After the completion of the study, participants were asked which footrest height they preferred before leaving the study area.

*Signal Processing:*

Ground reaction force and moment data (Fx, Fy, Fz, Mx, My, Mz) were sampled at 1000Hz using AMTI NetForce software and imported into custom MatLab script (MathWorks, Natick, MA, USA) for processing and analysis. Force data were filtered using a dual 10 Hz low-pass, 2$^{nd}$ order Butterworth filter and down sampled to 50 Hz for analysis (Mohan et al., 2014). The force data were then clipped into three, one minute bins for statistical analysis. The time bins included data from minutes 2.5-3.5, 7.5-8.5, and 12.5-13.5 which are referred to as the "start" "middle" and "end" phase, respectively. These time bins were chosen deliberately to preserve authentic data by avoiding the body shifts that were made while participants completed the discomfort questionnaires at minute 1, 5, 10 and 15 (Figure 4.2). It was assumed that the discomfort scores and force data were

collected in a temporally similar time frame and therefore were representative of each other. As such, the start, middle and end phase force data corresponded to the fifth, tenth and fifteenth minute discomfort scores.

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Insert Figure 4.2 Here

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Force calculations included percentage of weight distribution on either leg and COP range in both the A/P and M/L directions. VAS scores were measured with a ruler and converted into a score out of 100. The RPD scores required minimal processing and were collected for subjective analysis.

*Statistical Analysis:*

Dependent variables were averaged over each time window and entered into SPSS (Version 24.0) on Windows 10 (SPSS, Inc., Chicago, IL, USA) for statistical analysis. Pearson Correlation Coefficients were calculated for $COP_{A/P}$, $COP_{M/L}$ range and percentage weight distribution for both limbs. In situations where significant Pearson Correlations existed ($p < 0.05$, $r > 0.7$), multivariate ANOVAs were used to account for the shared variance. All other statistical analyses were completed using independent, mixed between-within repeated measures ANOVAs and statistical significance was set at $\alpha < 0.05$. In situations where the data did not meet sphericity, Greenhouse-Geisser corrected values were used. Significant main effects and interaction effects were further examined using Tukey's HSD post hoc test to identify where the effects existed.

## Results

Demographic information and data collected from outcome measures are displayed in Table 4.1. Descriptive statistics for the force data and discomfort scores are presented in Tables 4.2 and 4.3 respectively. Correlation coefficients and probability statistics for the force data are presented in Table 4.4 and the statistical results of the ANOVAs are found in table 4.5.

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Insert Table 4.1 – 4.5 here

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Significant Pearson correlation coefficients were observed for percentage weight distribution in both limbs ($r = -1.0$, $p < 0.001$) and for $COP_{range}$ between the A/P and M/L directions in the elevated limb ($r = 0.802$, $p < 0.001$) (Figures 4.3 and 4.4, respectively). As such, these variables were analyzed using a multivariate ANOVA procedure. All other dependent variables produced non-significant correlations and were therefore examined using univariate repeated measures ANOVAs.

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Insert figure 4.3 & 4.4 here

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*Force Data:*

The analysis of weight distribution demonstrated significant main effects of footrest height ($F(1,18) = 884.65$, $p < 0.001$, $\omega^2 = 0.986$) and leg ($F(1,13) = 5729.72$, $p < 0.001$, $\omega^2 = 0.998$) which drove the interaction effect between footrest height and leg ($F(1,39) = 884.65$, $p < 0.001$, $\omega^2 = 0.986$). During flat ground stance, weight was evenly distributed between the stance limb ($\bar{x} = 52.74\%$, SD = 3.52%) and elevated limb ($\bar{x} = 47.26\%$, SD = 3.52%). During footrest trials, significantly more weight was distributed to the stance limb ($\bar{x} = 85.38\%$, SD = 2.02%) compared to the elevated limb ($\bar{x} = 14.6\%$, SD = 2.02%) with no difference between footrest conditions (Figure 4.5). Weight distribution was significantly different ($p<0.01$) between the stance and elevated limb during footrest trials but not during flat ground stance.

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Insert Figure 4.5 here

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Significant main effects for footrest height ($F(3,39) = 11.857$, $p < 0.001$, $\omega^2 = 0.477$) and direction ($f(1,13) = 74.463$, $p < 0.001$, $\omega^2 = 0.851$) were observed between $COP_{A/P}$ and $COP_{M/L}$ range in the elevated limb (Figures 4.6 and 4.7, respectively). These main effects drove the interaction effect between COP range direction and footrest height ($F(3,39) = 3.263$, $p = 0.031$, $\omega^2 = 0.201$) in the elevated limb (Figure 4.8). As such, $COP_{A/P}$ and $COP_{M/L}$ ranges were both significantly larger when a footrest was present, with the medium and high footrest demonstrating significantly larger ranges than the low footrest (Figures 4.6 and 4.7, respectively). Furthermore, the COP range magnitude was significantly larger in the A/P direction than the M/L direction for all footrest conditions (Figure 4.8).

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Insert Figure 4.6 – 4.8 here

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A four-way interaction effect was observed between direction, footrest height, time and sex ($F(3,78) = 3.087$, $p = 0.009$, $\omega^2 = 0.192$) for $COP_{A/P}$ and $COP_{M/L}$ in the elevated limb (Figure 4.7). However, during post hoc testing, only one difference was observed out of the twenty-four comparisons made. The $COP_{A/P}$ range was significantly higher in males during the start of the low footrest trials than females. Otherwise, all other comparisons were non-significant.

*Discomfort Scores:*

A significant main effect of time was observed for low back discomfort ($F(1,15) = 4.65$, $p = 0.043$, $\omega^2 = 0.27$) such that lumbar discomfort increased steadily as a function of time regardless of standing condition (Figure 4.9).

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Insert Figure 4.9 here

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It is important to note a significant main effect of footrest height on lumbar discomfort was not observed ($F(2,23) = 3.4$, $p = 0.055$, $\omega^2 = 0.21$). Although it was trending towards significance, the relationship was barely non-significant.

There were no significant RPD findings and very few areas of discomfort were identified besides the low back. The most commonly identified areas included the hip, knee, calf and foot of the stance thigh. In fact, 10 of the 16 participants expressed discomfort in their foot. Unfortunately, the foot was not an option provided on the RPD form and was not objectively analyzed.

## Discussion

Consistent with our *a priori* hypothesis, the presence of a footrest affected the amount of weight that was distributed to either limb, with a significantly larger dependence on the stance limb. However, the amount of weight distributed to either leg was not dependent upon footrest height and was rather consistent across footrest conditions. By definition (Wang & Newell, 2014), all participants stood in a symmetrical stance position during flat ground stance with 52.76% +/- 3.6% ($\bar{x}$ +/- SD) of their body weight (BW) on the stance limb and 47.26% +/- 3.6% ($\bar{x}$ +/- SD) on the elevated limb. Conversely, when using a footrest, individuals stood in an asymmetrical stance position with 85.38% +/- 17% ($\bar{x}$ +/- SD) of their BW on the stance limb and 14.6% +/- 17% ($\bar{x}$ +/- SD) on the elevated limb. Previous research has reported an approximate 80%/20% split between stance and elevated limbs when using a footrest (Mohan et al., 2014). Our findings suggest a slightly stronger dependence on the stance limb with an approximate 85%/15% weight distribution. The observed reduction in weight distribution on the elevated limb was echoed by findings in a parallel study that found a reduction in hip muscle activity in the elevated limb (Chapter 3). Gluteus medius (GM) activation was significantly lower in the elevated limb during footrest trials when compared to flat ground stance. The observed decrease in gluteus medius activity suggests a reduced need for local stabilization in the hip to accommodate for less weight bearing with the elevated limb.

COP range was larger in both the A/P and M/L direction in the elevated limb when a footrest was present. The magnitude of COP increased sequentially as a function of footrest height and COP range was always larger in the A/P direction. As the height of the footrest increased, the body was placed in a more unstable and overtly asymmetrical position compared to flat ground stance. Previous research has shown that footrest usage results in altered lumbo-pelvic and hip joint angles (Fewster et al., 2017; Chapter 3). The posture associated with footrest usage results in joint angle changes and muscle length differences which manifest as a novel proprioceptive environment for the sensory system. These proprioceptive differences are a result of altered feedback from the joint capsules, golgi tendon organs (GTO's) and muscle spindles which help to govern balance (Kistemaker et al., 2012). The processing pathways and motor control strategies that individuals adopt while using a footrest are not as well established as flat ground stance which can lead to overcorrection and increased sway profiles. Mohan and colleagues (2015) demonstrated this when they observed increased COP mean distance, RMS, range, and velocity in both the A/P and M/L direction when a footrest was being used.

In the current study, the observed COP range differences were not evident in the stance limb. This likely reflects the different roles that each limb plays in maintaining stability during asymmetrical postures. The hind limb acts as a stable base of support with 85% of an individual's body weight whereas the forelimb acts as a mobilizing limb that can easily adjust its activity to respond to perturbation. A small increase in motor unit firing in the elevated limb has a larger effect on COP compared to the same change in muscle activity in the stance limb because it takes more force to move a larger load (Chapter 3). Additionally, the GM on the elevated limb has a higher potential motor recruitment ability because of its lower baseline activity in comparison to the elevated limb. Therefore, individuals use the elevated limb to make minor adjustments to monitor balance on a micro-level while using a footrest. This helps to minimize energy consumption and improves efficiency by recruiting smaller motor units to make minor postural adjustments rather than large overcorrections by tonic muscles in the stance limb.

In agreement with previous research, we found a strong correlation between sway in the frontal and sagittal plane (Mohan et al., 2014). In other words, when sway increased in one plane, it also increased in the perpendicular direction. This pattern suggests that multiple motor unit strategies are at play when maintaining balance on a footrest (Winter et al., 1996). The inverted pendulum model of balance distinguished between a bottom-up, ankle strategy (plantar flexion and dorsiflexion) that governs A/P excursion and a top-down hip strategy (adduction and abduction) that governs M/L sway (Winter, 1993). However, these strategies only relate to side-by-side quiet stance and not asymmetrical postures. Subsequent research observed that an intermediate 45° staggered stance involved both ankle and hip strategies in a complex manner to control net balance (Winter, 1996). In the M/L direction, hip and ankle strategies worked together in a summative fashion. In the A/P direction, the hips are the main driver of sway and ankle activity works in a reductionist manner to cancel out inappropriate contributions from the hips. In this way, the ankles act as an inhibitory mechanism to prevent overcorrection in staggered standing positions (Winter, 1996).

Previous research has shown that footrests are the most effective workplace standing aid at altering posture compared to flat ground stance (Fewster et al., 2017). Footrests have also been shown to significantly affect muscle activation in the lumbar spine and gluteal region on the elevated limb side (Chapter 3). In addition to objective biomechanical factors, it is important to consider discomfort as a potentially important indicator of user compliance and implementation rates. Previous research suggests that individuals will use a footrest 83% of the time when one is provided to them during a 2 hour standing period (Rys & Konz, 1994). In a healthy population, this study found that individuals experienced significant increases in discomfort over time regardless if a footrest was provided to them or not. This effect was time dependent, and resulted in the most discomfort towards the end of each trial. However, a significant difference in discomfort between footrest conditions or in comparison to flat ground stance was not observed. This suggests that footrests are not significantly more uncomfortable than flat ground stance during a 15 minute standing trial. However, the results were strongly

trending towards statistical significant with increasing discomfort as each trial progressed. With the use of a larger sample, such significance would likely have been observed.

Based upon the VAS scores, the flat ground trial was the most comfortable ($\bar{x}$ = 13.7 mm, SD = 3.5) followed by the tallest footrest ($\bar{x}$ = 21.2mm , SD = 4.2), the low footrest ($\bar{x}$ = 27.8 mm, SD = 4.1) and then the medium footrest ($\bar{x}$ = 29.6 mm, SD = 4.8). Although non-significant, the difference between flat ground and the medium and high footrest surpass the minimum clinically significant difference of 10 mm and may be worth reporting due to its potential clinical significance. This difference would likely become more apparent with a larger sample size.

Contrary to the VAS scores, 46.67% of participants identified the medium footrest as the most comfortable footrest condition, 33.33% preferred the low footrest and 20% preferred the tallest footrest following the study. The medium footrest (20 cm) equated to approximately 11.6% of the participant body height which closely matches the recommendation of 10% body height for individuals suffering from non-specific LBP (Son et al., 2015). It also falls within the current recommendations to use a footrest measuring between 10-20 cm in height (Ebben,2003). However, participants retrospective preference did not correspond well to their VAS scores reported in real time.

An unexpected observation from the RPD scale showed that 67% of participants identified the stance foot as a pain generator at some point in the study. This was not foreseen prior to the study and was therefore not examined appropriately. Stance foot discomfort may be a major concern while using a footrest and the relationship should be investigated further by expanding the RPD score to include the foot location, or recruiting a foot specific outcome measure.

Our force platform design provided a novel approach to studying balance during asymmetrical postures. However, the difference in elevation between force platforms during footrest trials rendered the COPnet calculations inaccurate. Therefore, COPnet calculations were not calculated or considered in this analysis. Methodological limitations

include a lack of left foot dominant participants and a potential order effect having flat ground stance as the first trial. Future studies should aim to examine how switching between feet affects comfort levels because it reflects a likely scenario that would occur in the workplace. Additionally, future analyses should focus on how footrest usage affects comfort in the lower limb including the ankle and foot. Additionally, administering the discomfort questionnaires on the computer would provide a viable approach to preventing acute postural shifts during each trial.

The results of this study suggest that utilization of a footrest results in altered weight distribution and increased magnitude of sway in the elevated limb in comparison to flat ground stance. Weight distribution was not affected by changes in footrest height ranging from 10-30 cm. However, the magnitude of sway in the elevated limb increased as the height of the footrest increased. Therefore, the use of a footrest does indeed affect weight distribution and sway based upon COP range magnitude. However, the goal of this study was to compare various footrest heights and potentially recommend one height over the others. Based upon the results, we are unable to firmly recommend the preferred usage of one footrest height over the other. The lowest footrest caused the smallest change in sway in both the medial and lateral direction. The medium footrest was the most popular footrest but did not provide benefits over the low or tall footrest in any objective measures. Finally, the tallest footrest demonstrated the least discomfort but the largest sway profile in the M/L direction.

In conclusion, the findings of this study support the usage of footrests but provide no firm recommendation as to what the ultimate height of that footrest should be. From a psychological perspective, providing workers with the autonomy to choose their preferred footrest within the 10-20 cm range may empower them to incorporate a footrest into their daily routine and improve compliance rates.

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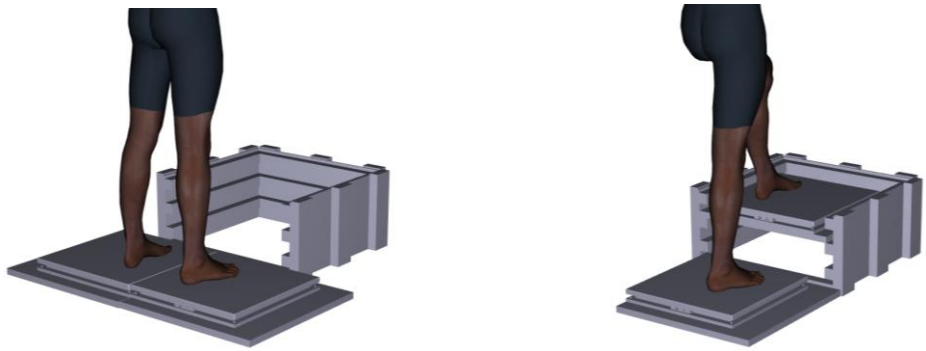

***Figure 4.1: Force platform design during the flat ground trial (left) and during the medium footrest trial (right). The platform was inserted into the grooves that were specifically measured to allow 10 cm, 20 cm and 30 cm between each force platform.***

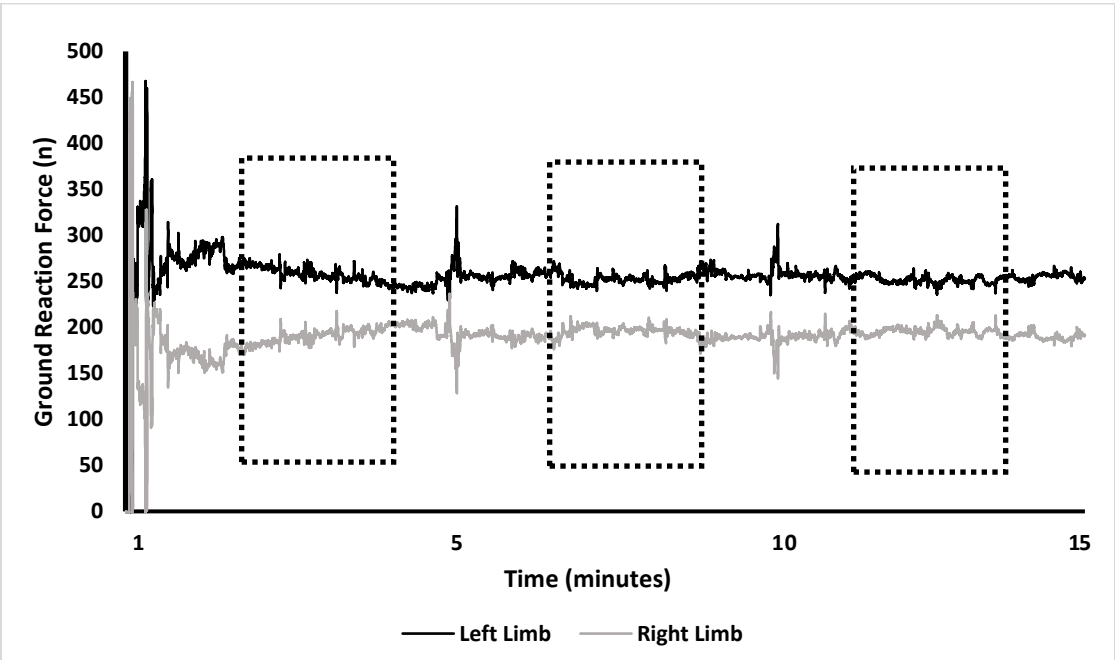


***Figure 4.2: Three time windows of force data were used for analysis (dotted lined boxes). Note the spikes in ground reaction forces at the one minute mark while mounting the platform and at the five and ten minute mark while individuals completed the discomfort questionnaires. This data represents a typical flat ground trial.***

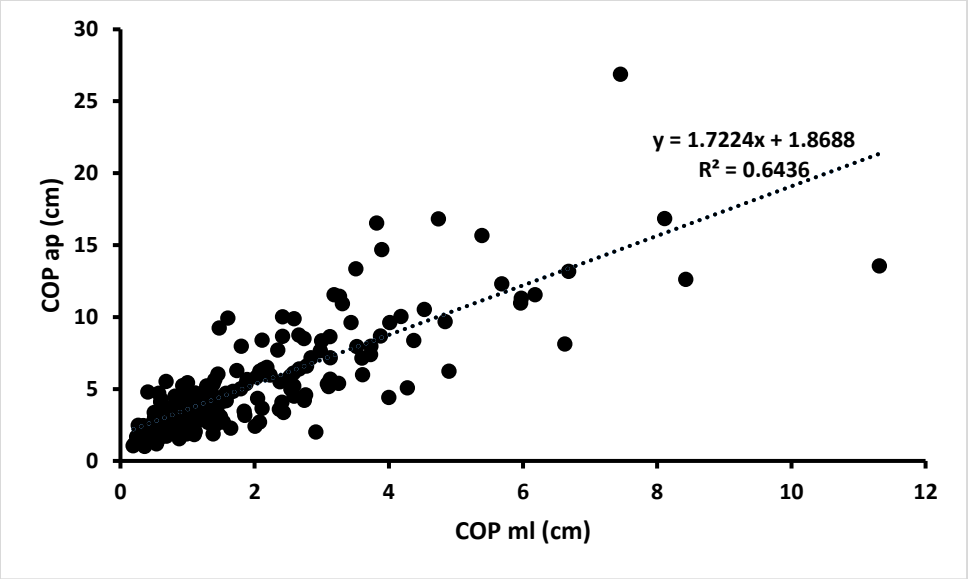


***Figure 4.3: Pearson's correlation representing the moderately positive relationship between $COP_{ap}$ and $COP_{ml}$ range in the elevated limb.***

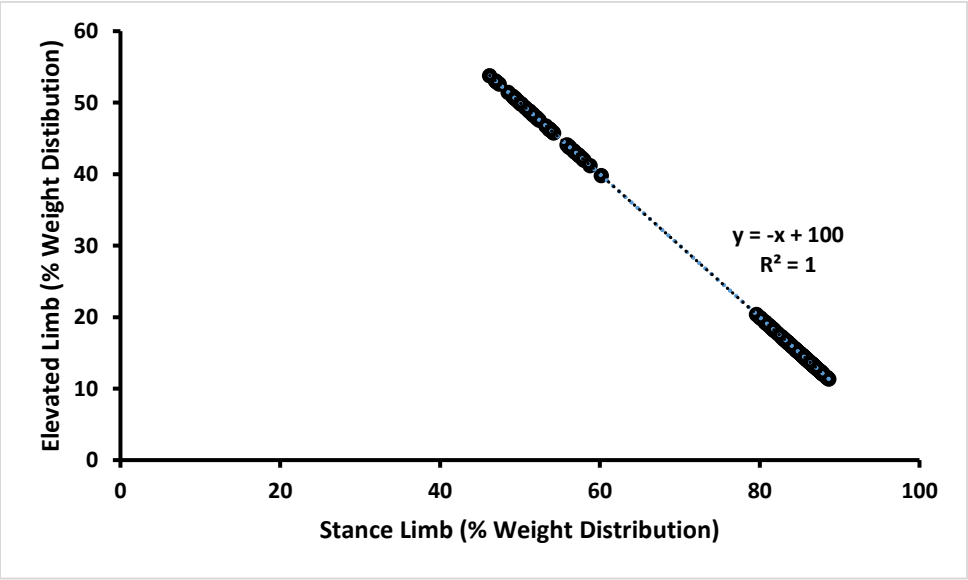


***Figure 4.4: Pearson's correlation for weight distribution between the stance and elevated limb indicating a strong negative relationship.***

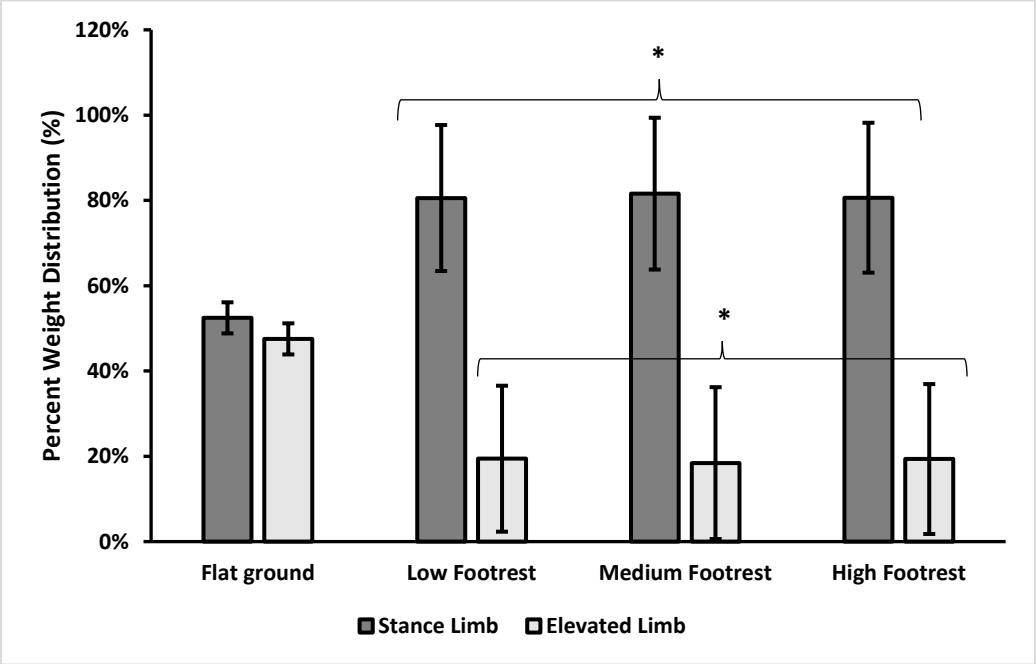


***Figure 4.5: A significant interaction effect between footrest height and leg for Percentage Weight Distribution. The asterisk (*) represents significantly different values when compared to flat ground stance. Significance was also seen between the stance and elevated limb during footrest trials. There was no significant difference between footrest heights for either limb.***

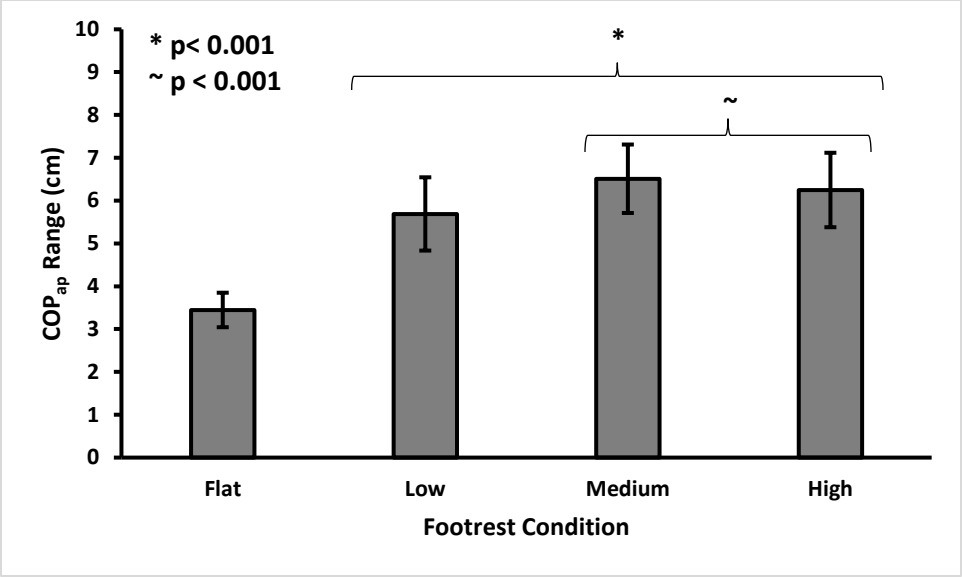


***Figure 4.6: $COP_{ap}$ Range for the elevated limb. The asterisk (*) represents significantly larger ranges than flat ground stance. The tilde (~) representes significantly larger range than the low footrest.***

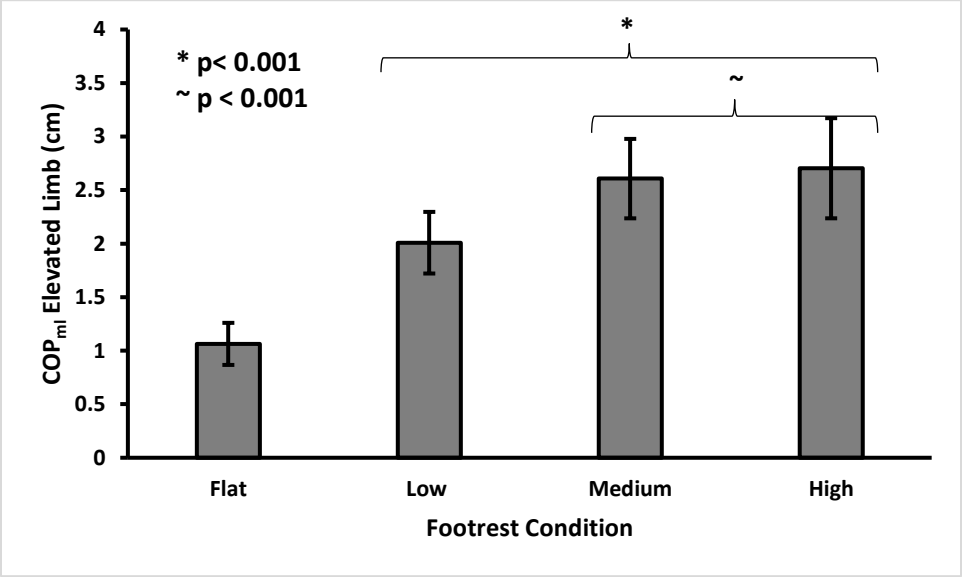


***Figure 4.7: $COP_{M/L}$ Range for the elevated limb. The asterisk (*) represents significantly larger range than flat ground stance. The tilde (~) representes significantly larger range than the low footrest.***

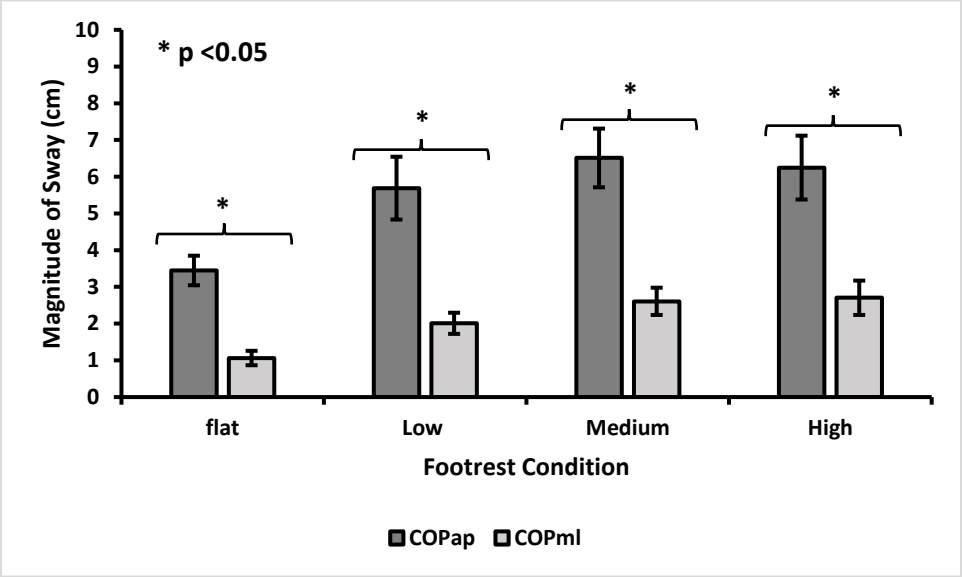


***Figure 4.8: A two-way interaction effect between footrest condition and COP range direction ($COP_{A/P}$ vs $COP_{M/L}$). The asterisk (*) represents significant differences between $COP_{A/P}$ and $COP_{M/L}$.***

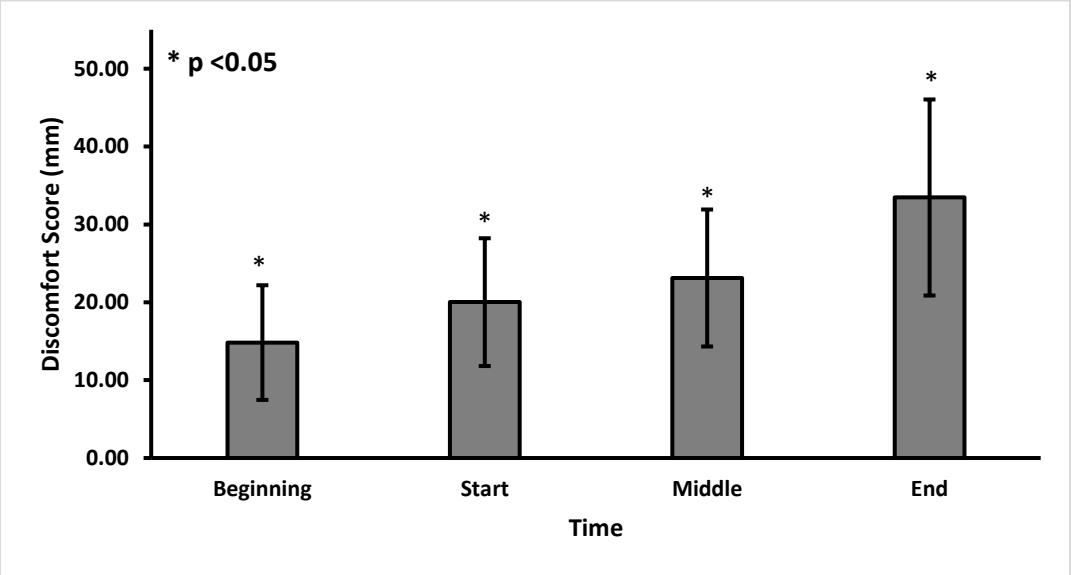


***Figure 4.9: Discomfort scores as a function of time. The asterisk (*) represents gradually increasing significant differences between time periods regardless of footrest condition.***

**Table 4.1: Demographic Information of Male and Female Participants including Means ($\bar{x}$) and Standard Deviations (SD) of Age, Height, Weight and Written Outcome Measure Scores.**

| | Male: | | Female: | | Total: | |
|---|---|---|---|---|---|---|
| | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD |
| Age: | 25.29 | 4.35 | 23.38 | 4.75 | 24.27 | 4.51 |
| Height: | 181.43 | 5.06 | 165.81 | 7.88 | 173.10 | 10.35 |
| Weight: | 72.55 | 12.83 | 60.12 | 10.49 | 173.10 | 16.25 |
| WHQ Score: | 29.57 | 35.94 | 56.88 | 14.02 | 44.13 | 29.17 |
| WFQ Score: | 4.86 | 9.21 | 11.13 | 8.20 | 8.20 | 8.97 |
| VAS: | 0.57 | 0.98 | 0.38 | 0.74 | 0.36 | 0.83 |
| OBDI Score: | 1.57 | 1.72 | 1.63 | 3.29 | 1.60 | 2.59 |

**Table 4.2: Mean ($\bar{x}$) and Standard Deviation (SD) of Force Data Separated by Limb, Footrest Condition, Time and Measure. Measures include Percentage of Weight Distribution, COP range in the anterior/posterior (A/P) and medial/lateral (M/L) Directions.**

| | | % Weight Distribution | | | | $COP_{A/P}$ range | | | | $COP_{M/L}$ range | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | Stance Limb | | Elevated Limb | | Stance Limb | | Elevated Limb | | Stance Limb | | Elevated Limb | |
| | | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD |
| **Flat Ground** | **Start** | 0.53 | 0.03 | 0.47 | 0.03 | 0.03 | 0.03 | 0.03 | 0.02 | 0.01 | 0.01 | 0.01 | 0.02 |
| | **Middle** | 0.52 | 0.04 | 0.48 | 0.04 | 0.04 | 0.03 | 0.03 | 0.02 | 0.01 | 0.01 | 0.01 | 0.01 |
| | **End** | 0.53 | 0.04 | 0.47 | 0.04 | 0.04 | 0.03 | 0.04 | 0.02 | 0.01 | 0.02 | 0.01 | 0.01 |
| | **Full** | 0.53 | 0.04 | 0.47 | 0.04 | 0.04 | 0.03 | 0.03 | 0.02 | 0.01 | 0.01 | 0.01 | 0.01 |
| **Low Footrest** | **Start** | 0.85 | 0.02 | 0.15 | 0.02 | 0.03 | 0.02 | 0.06 | 0.07 | 0.01 | 0.01 | 0.02 | 0.02 |
| | **Middle** | 0.85 | 0.02 | 0.15 | 0.02 | 0.03 | 0.03 | 0.04 | 0.02 | 0.01 | 0.01 | 0.02 | 0.01 |
| | **End** | 0.85 | 0.03 | 0.15 | 0.03 | 0.05 | 0.03 | 0.06 | 0.04 | 0.02 | 0.01 | 0.02 | 0.02 |
| | **Full** | 0.85 | 0.02 | 0.15 | 0.02 | 0.04 | 0.03 | 0.06 | 0.04 | 0.01 | 0.01 | 0.02 | 0.02 |
| **Medium Footrest** | **Start** | 0.86 | 0.02 | 0.14 | 0.02 | 0.03 | 0.02 | 0.07 | 0.04 | 0.01 | 0.01 | 0.02 | 0.02 |
| | **Middle** | 0.86 | 0.01 | 0.14 | 0.01 | 0.03 | 0.03 | 0.06 | 0.03 | 0.01 | 0.01 | 0.02 | 0.02 |
| | **End** | 0.86 | 0.01 | 0.14 | 0.01 | 0.04 | 0.04 | 0.07 | 0.03 | 0.01 | 0.01 | 0.03 | 0.02 |
| | **Full** | 0.86 | 0.01 | 0.14 | 0.01 | 0.03 | 0.03 | 0.06 | 0.03 | 0.01 | 0.01 | 0.03 | 0.02 |
| **High Footrest** | **Start** | 0.85 | 0.02 | 0.15 | 0.02 | 0.03 | 0.02 | 0.06 | 0.04 | 0.01 | 0.01 | 0.03 | 0.03 |
| | **Middle** | 0.85 | 0.02 | 0.15 | 0.02 | 0.04 | 0.04 | 0.06 | 0.04 | 0.01 | 0.01 | 0.03 | 0.02 |
| | **End** | 0.85 | 0.02 | 0.15 | 0.02 | 0.05 | 0.04 | 0.06 | 0.03 | 0.02 | 0.01 | 0.03 | 0.02 |
| | **Full** | 0.85 | 0.02 | 0.15 | 0.02 | 0.04 | 0.03 | 0.06 | 0.04 | 0.01 | 0.01 | 0.03 | 0.02 |

**Table 4.3: Mean ($\bar{x}$) and Standard Deviation (SD) of Visual Analogue Scale (VAS) Responses Separated by Footrest Condition and Time.**

| | | Male: | | Female: | | Both: | |
|---|---|---|---|---|---|---|---|
| | | $\bar{x}$ | SD | $\bar{x}$ | SD | $\bar{x}$ | SD |
| Flat ground | Pre-Trial | 0.00 | 0.00 | 0.14 | 0.22 | 0.08 | 0.17 |
| | Start | 0.00 | 0.00 | 0.25 | 0.42 | 0.13 | 0.32 |
| | Middle | 0.00 | 0.00 | 0.28 | 0.42 | 0.15 | 0.33 |
| | End | 0.00 | 0.00 | 0.42 | 0.66 | 0.22 | 0.51 |
| Low | Pre-Trial | 0.12 | 0.32 | 0.25 | 0.35 | 0.19 | 0.33 |
| | Start | 0.20 | 0.25 | 0.30 | 0.47 | 0.25 | 0.37 |
| | Middle | 0.18 | 0.31 | 0.41 | 0.44 | 0.30 | 0.39 |
| | End | 0.27 | 0.50 | 0.50 | 0.58 | 0.39 | 0.53 |
| Medium | Pre-Trial | 0.00 | 0.00 | 0.34 | 0.49 | 0.18 | 0.39 |
| | Start | 0.06 | 0.16 | 0.46 | 0.52 | 0.27 | 0.43 |
| | Middle | 0.14 | 0.28 | 0.43 | 0.50 | 0.29 | 0.43 |
| | End | 0.33 | 0.56 | 0.61 | 0.68 | 0.48 | 0.62 |
| High | Pre-Trial | 0.00 | 0.00 | 0.33 | 0.49 | 0.18 | 0.39 |
| | Start | 0.03 | 0.08 | 0.30 | 0.49 | 0.18 | 0.38 |
| | Middle | 0.06 | 0.16 | 0.36 | 0.57 | 0.22 | 0.44 |
| | End | 0.18 | 0.48 | 0.37 | 0.55 | 0.28 | 0.51 |

**Table 4.4: Pearson's Correlation Coefficients (r) for COP Range and Percentage of Weight Distribution Within and Between Limbs. The Asterisk (*) Represents Significant Findings (P<0.05) and a Double Asterisk (**) represents Clinically Significant Relationships (r > +/- 0.7).**

| | | | Stance Limb | | | Elevated limb | | |
|---|---|---|---|---|---|---|---|---|
| | | | **$COP_{A/P}$ range** | **$COP_{ml}$ range** | **Distribution** | **$COP_{A/P}$ range** | **$COP_{ml}$ range** | **Distribution** |
| **Stance Limb** | **$COP_{A/P}$ range** | **r** | | 0.692 | -0.00741683 | 0.518 | 0.528 | 0.00741683 |
| | | **p** | | <0.001* | 0.921284895 | <0.001* | <0.001* | 0.921284895 |
| | **$COPml_{range}$** | **r** | | | 0.005595086 | 0.359 | 0.443 | -0.005595086 |
| | | **p** | | | 0.940577871 | <0.001* | <0.001* | 0.940577871 |
| | **Distribution** | **r** | | | | 0.304 | 0.33 | **-1.000**** |
| | | **p** | | | | <0.001 | <0.001 | **<0.001*** |
| **Elevated limb** | **$COP_{A/P}$ range** | **r** | | | | | **.802**** | -0.304 |
| | | **p** | | | | | **<0.001*** | <0.001 |
| | **$COPml_{range}$** | **r** | | | | | | -0.330 |
| | | **p** | | | | | | <0.001 |
| | **Distribution** | **r** | | | | | | |
| | | **p** | | | | | | |

**Table 4.5: Main Effects and Interaction Effects Resulting from the ANOVA's Investigating Percentage of Weight Distribution and COPrange. The Asterisk Denotes Significance.**

| | | Stance Limb | | | | | | Elevated Limb | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | $COP_{A/P}$ range | | $COP_{M/L}$ range | | % Weight Distribution | | $COP_{A/P}$ range | | $COP_{M/L}$ range | | % Weight Distribution | |
| | | F | Sig | F | Sig | F | Sig | F | Sig | F | Sig | F | Sig |
| Main Effects | Condition | 0.97 | 0.39 | 0.74 | 0.54 | **884.65** | **<0.01*** | **8.74** | **<0.01*** | **13.94** | **<0.01*** | **884.65** | **<0.01*** |
| | Time | **3.58** | **0.042*** | 2.96 | 0.07 | 0.59 | 0.56 | 1.92 | 0.17 | 1.57 | 0.23 | 0.59 | 0.56 |
| | Leg | | | | | **5729.71** | **<0.01*** | | | | | **5729.71** | **<0.01*** |
| | Sex | 0.15 | 0.71 | 3.83 | 0.07 | 0.29 | 0.60 | 0.74 | 0.41 | 1.46 | 0.25 | 0.29 | 0.60 |
| | Direction | | | | | | | **74.46** | **<0.01*** | **74.46** | **<0.01*** | | |
| 2- Way Interactions | Condition* Time | 0.60 | 0.73 | 0.30 | 0.84 | 1.01 | 0.42 | 1.05 | 0.38 | 0.70 | 0.65 | 1.01 | 0.42 |
| | Condition * Leg | | | | | **884.63** | **<0.001*** | | | | | **884.63** | **<0.01*** |
| | Condition*Sex | 0.02 | 0.98 | 0.50 | 0.68 | 0.19 | 0.90 | 2.04 | 0.12 | 0.24 | 0.87 | 0.19 | 0.90 |
| | Condition* Direction | | | | | | | **3.26** | **0.03*** | **3.26** | **0.03*** | | |
| | Sex*Leg | 0.29 | 0.60 | N/A | N/A | 0.29 | 0.60 | | | | | 0.29 | 0.60 |
| | Sex* Time | 1.48 | 0.25 | 1.12 | 0.34 | 2.57 | 0.10 | 1.24 | 0.31 | 2.91 | 0.07 | 2.57 | 0.10 |
| | Sex*Direction | | | | | | | 0.23 | 0.64 | 0.23 | 0.64 | | |
| | Time*Leg | | | | | 0.59 | 0.56 | | | | | 0.59 | 0.56 |
| | Time* Direction | | | | | | | 1.64 | 0.21 | 1.64 | 0.21 | | |
| 3 and 4-Way Interactions | Time*Direction*Sex | | | | | | | 0.31 | 0.72 | 0.31 | 0.72 | | |
| | Time*Direction*Condition | | | | | | | 1.78 | 0.17 | 1.78 | 0.17 | | |
| | Time*Condition*sex | 0.80 | 0.58 | 0.55 | 0.77 | 0.49 | 0.82 | 1.91 | 0.09 | 0.90 | 0.50 | 0.49 | 0.82 |
| | Time*Condition*leg | | | | | 1.01 | 0.42 | | | | | 1.01 | 0.42 |
| | Time*Condition*Leg*Sex | | | | | 0.49 | 0.82 | | | | | 0.49 | 0.82 |
| | Direction*Condition*Sex | | | | | | | 2.35 | 0.09 | 2.35 | 0.09 | | |
| 4-way | Time*Condition*Leg*Sex*Direction | | | | | | | **3.09** | **0.04*** | **3.09** | **0.04*** | | |

# CHAPTER 5: SUMMARY AND CONCLUSIONS

**Andrew C. Cregg & Lori A Livingston**

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

Corresponding Author

Lori A. Livingston, PhD

Faculty of Health Sciences

University of Ontario Institute of Technology (UOIT)

2000 Simcoe St. N.

Oshawa, ON, Canada

L1H 7K4

Telephone: (905)721-8668

Email: Lori.Livingston@uoit.ca

This chapter aims to summarize the findings that were expressed throughout this thesis and to synthesize an over-arching conclusion that can help to guide future recommendations. It begins by revisiting the rationale for the study and the overall purpose followed by a reiteration of the findings from each chapter and how they contribute towards the overall conclusion and recommendations on how to update the current guidelines. Finally, the chapter will suggest future directions of research for this line of inquiry.

## Purpose and Rationale

The purpose of this study was to examine how various footrest heights affect joint angles, muscle activation, centre of pressure (COP) and discomfort in comparison to flat ground stance. The study included a 10 cm and 20 cm footrest to represent the current recommendations, and a 30 cm footrest to represent the tallest commercially available footrest. This study was heavily influenced by recent research that focused on various standing aids and their effect on the human body (Dolan et al., 1988; Ebben et al., 2003; Fewster et al., 2017; Gallagher 2014; Gallagher & Callaghan, 2016; Lee et al., 2018; Mohan et al., 2014; Son et al., 2017). Fewster and colleagues (2017) compared four different standing aids through assessment of joint angles and electromyography (EMG). They concluded that the footrest usage was the only effective intervention at inducing lumbar spine flexion. Mohan and colleagues (2014) examined how footrest usage affects postural stability and centre of pressure (COP) using force platforms and found that footrest usage resulted in negative effects on postural sway. However, both of these studies examined very short standing periods of five minutes and 30 seconds respectively.

Only one previous study has examined the effect of footrest heights on muscle fatigue, kinematics and kinetics (Son et al., 2017). This study used normalized footrests (5%, 10% and 15% of body height) and concluded that a 10% body height footrest was most appropriate because it minimized fatigue, external lumbar moments and discomfort. However, this study was conducted on a clinical population involving those with non-specific low back pain (LBP) and cannot be properly applied to a healthy population. Only one study has reported on the negative effects associated with footrest usage. Lee and colleagues (2018) reported that footrest usage

was actually associated with LBP development in healthy individuals over a two hour standing period with and without a footrest.

Given that only one previous study has attempted to systematically examine footrest height, this became the primary focus of this study. Footrest height recommendations are poorly defined and warrant further examination. Therefore, this study incorporated commonly utilized measures including kinematics, EMG, force and discomfort scores to examine this relationship. To provide unique contributions to the literature, this study utilized 15 minute standing periods, examination of the lumbar-to-thigh angles, EMG assessment of the tensor fascia lata (TFL), a dual force plate design and administration of discomfort questionnaires every five minutes to a healthy population rather than a clinical one.

Previous research has identified short trials (0.5 – 5 minute trials) as a major limitation (Fewster et al., 2017). Conversely, trials lasting 2 hours far surpass the recommended upper limit of 40 minutes for prolonged standing that was suggested by Coenen and colleagues (2017). As such, trials were set at 15 minutes to represent a likely amount of time that a worker would stand in any one position. The TFL is considered one of the main stabilizers of the hip and has not been previously considered in similar analyses (Flack et al., 2012). The dual force platform design provides a unique methodological feature that has not been implemented in previous footrest studies and allows us to examine COP under each limb separately.

## Key Findings

Chapters three and four were separated for readability and to explain each dependent variable without overlooking or underrepresenting any significant findings. It is important to combine these findings into a coherent synthesis to examine the overarching results as a whole. Use of a footrest was associated with increased flexion through the thoraco-sacral, lumbo-sacral and lumbar-to-thigh angles on the elevated limb. The amount of flexion was dependent upon footrest height up to 20 cm, but decreased when the 30 cm footrest was used. Over time, lumbo-sacral flexion increased regardless of footrest condition and males stood with more flexion overall. When comparing the lumbar-to-thigh angles between limbs, it became apparent that flexion in the elevated limb was offset by extension in the stance limb to maintain a consistent

net hip angle. In summary, a footrest induced spinal flexion, elevated hip flexion and equivalent stance limb hip extension. Joint angles were dependent upon footrest height up to 20 cm and then a slight reduction above that limit.

Muscle activity was also affected by footrest usage but the effect was isolated to the elevated limb. When individuals utilized a footrest, they demonstrated reduced gluteus medius (GM) activity and elevated lumbar erector spinae (LES) activity on the side of the elevated limb. The reduction in GM activity is theorized to reflect a reduction in stabilization requirements for the elevated limb. The elevated LES activity reflects an increased need for lumbar stabilization to support the elevated limb. Although non-significant, GM activity was elevated on the stance limb and was trending towards significance.

While standing on flat ground, participants maintained symmetrical stance with approximately 50% of their weight distributed to either limb. During footrest trials, dependence was shifted to the stance limb which held approximately 85% of the body weight while the elevated limb held the remaining 15%. As such, the elevated limb requires less forceful stabilization which was reflected in the GM activity. Secondly, COP range in the medial/lateral (M/L) and anterior/posterior (A/P) directions were significantly higher when a footrest was in use. This effect was likely observed in the elevated limb because it's responsible for less weight management. As such, smaller changes in COP force magnitude and direction will have a larger effect on the COP. Additionally, the elevated limb is hypothesized to be respond to minor perturbations because the muscular system has a lower baseline activity and therefore more potential motor recruitment availability. The asymmetrical stance position associated with footrest usage results in a novel posture that alters proprioceptive feedback and motor pattern responding resulting in increased sway.

Footrest usage did not cause increases in discomfort over the 15 minute trial. However, it did meet minimal clinically significant values indicating discomfort and was trending towards significance. The flat ground trial was the most comfortable based upon VAS scores ($\bar{x}$ = 13.7 mm, SD = 3.5) followed by the tallest footrest ($\bar{x}$ = 21.2mm , SD = 4.2), the low footrest ($\bar{x}$ = 27.8 mm, SD = 4.1) and then the medium footrest ($\bar{x}$ = 29.6 mm, SD = 4.8). However, based on individual

preference, 46.67% of participants identified the medium footrest as the most comfortable footrest. This equates to approximately 11.6% of our samples average height and compares well with Son and colleagues (2017) recommendation.

The goal of this thesis was to improve the current footrest height recommendations. The intention was to investigate whether one single footrest recommendation could be applied to the working population as a whole. The rationale behind this investigation was to clarify the currently vague footrest height recommendation to improve implementation procedures and compliance within the workplace. In isolation from each other, neither of the described studies provides adequate evidence to make a single footrest recommendation. They both support the current recommendation that includes footrests within a 10-20 cm range. However, when the findings are considered together, a stronger recommendation can be made by considering the positive and negative effects that each footrest has on the body.

## Final Recommendations

The results of the study conclude that utilization of a footrest was associated with numerous positive benefits regardless of its height. To narrow down a single recommendation, we must attempt to identify a footrests that provides unique advantages that are not observed with other footrests. When comparing strengths and weaknesses, it becomes clear that the 20 cm footrest should be recommended over the 10 cm and 30 cm alternatives. The 20 cm footrest induced the largest flexion angle through the thoraco-sacral and lumbo-sacral areas, caused the largest change in extension and flexion in the stance and elevated limbs respectively and induced the least change in ES activity. Additionally, it was rated as the most comfortable footrest. Twenty centimeters also matches well with Son and colleagues (2017) recommendation that individuals should use a normalized 10% body height footrest. Negative factors associated with the 20 cm footrests include that it induced the largest change in COP range and participants reported the highest VAS score with that footrest. Despite these negative factors, it was the only footrest to show significant, objective advantages over the other two footrests.

## General Limitations and Future Directions

This study should have included a kinematic analysis of individuals while they remained seated so that a comparison could be made between standing, standing with a footrest and sitting. This would have strengthened the argument for gradual postural change when individuals transition from a seated posture to using a footrest and eventually to a seated posture. These results could then have been compared to the recent findings by Glinka and colleagues (2018) who utilized a transitional chair to examine the postural continuum. Future kinematic studies should incorporate a flexion/extension – lateral bending rotation sequence to account for multi-axis motion about the spine rather than examining strict sagittal plan motion (Fewster et al., 2017). This may account for some of the variability observed in the spinal joint angles in the current study. The results of the study indicated that the stance limb was a significant contributor towards discomfort. Future studies should focus on lower limb discomfort and quantifying posture and muscle activation associated with each position change.

Another limitation of this study was the sample size. Although n = 16 was sufficient according to the sample size calculation, it was minimal which increases the possibility of Type 1 errors. In an ideal situation, a larger sample size would have been preferred. For data contamination purposes, the second manuscript only includes 15 participants which may distort the results slightly and increases the risk of Type 2 errors. Future studies should focus on examining if an appropriate dose-response relationship exists while using a footrest. Comparison studies should examine how elevating the dominant foot compares to elevating the non-dominant foot and could focus on examining how left and right side dominant individuals respond while standing with a footrest.

Given that a significant amount of literature has documented the physiological differences that separate PDs and NPDs, this line of inquiry could be expanded to include this classification system to understand how each group responds to various footrest heights. Previous literature has shown that PDs benefit from footrest usage but NPDs may be negatively influenced (Fewster et al., 2017). It is currently not known whether this effect is dependent upon the height of the footrest or not. Finally, future research should focus on workplace

implementation studies to examine the efficacy associated with using various footrests in the workplace. Laboratory studies give us an excellent understanding of the objective component parts, but implementation studies shed light on how these interventions fair in the real world.

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## APPENDIX A: INFORMED CONSENT

***Title: Changes in lumbar spine posture, muscle activation and centre of pressure while standing with different footrest heights during a standardized computer task.*** *This study has been approved by the UOIT Research Ethics Board REB #14477 on August 14th, 2017.*

**Principal Investigator:**

Dr. Lori Livingston
Dean, Faculty of Health Sciences
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4
Phone #: 905.721.8668
Email: lori.livingston@uoit.ca

**Student Lead:**

Andrew Cregg
Graduate Student, Faculty of Health Sciences
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4
Phone #: 647.285.4878
Email: Andrew.cregg@uoit.ca

**Supervisory Committee:**

Dr. Paul Yielder
Professor, Faculty of Health Sciences
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4
Phone #: 905.721.8668 ext. 2768
Email: paul.yielder@uoit.ca

Dr. Nicholas La Delfa
Assistant Professor
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4
Phone #: 905.721.8668 ext. 2139
Email: nicholas.ladelfa@uoit.ca

**Rationale:**

Research indicates that workers spend up to 75% of their workday in a prolonged seated posture. Maintaining a seated position for a prolonged period of time has been associated with development of low back pain. In fact, 84% of the working population will experience low back pain at some point in their lives. This overwhelming association between prolonged sitting and low back pain has lead ergonomic and medical experts to recommend that workers should incorporate some form of standing throughout the day. However, further research has indicated that prolonged standing can also lead to low back pain. Therefore, both seated and standing postures should be incorporated within the workplace.

Recommendations are constantly changing to represent the current understanding within the research. As such, the recommendations around sit-stand desks have been dramatically improved upon over the years. Industry leaders now recommend that workers should utilize a footrest to rest one foot on while standing to prevent low back pain. However, no research exists that investigates how the footrest height might affect our body mechanics. This research aims to improve these recommendations and to identify the most appropriate footrest height for workers. The study intends to investigate how footrests of different heights affect body mechanics.

**Purpose:**

The purpose of this study is to investigate how using footrests of different heights affects low back posture, muscle activation and postural stability in young healthy adults. This research aims to contribute one small piece to the larger ergonomic field to improve industry recommendations.

**Information for Participants:**

We are seeking healthy participants between 18 and 45 years of age. We are looking for participants who do not have a history of low back pain severe enough to have sought medical intervention or taken more than 3 days off work in the past 12 months. Participants must not have worked a job that requires prolonged standing over the last 12 months. Participants must also not have had hip or lumbar surgery and they must be able to stand for two hours consecutively.

***We encourage you to read this form thoroughly and ask any questions that you may have.*** Your participation in this study is entirely voluntary (your choice), and you are free to decline taking part in this study. If you agree to participate, you may withdraw from the study at any time without giving a reason. This will in no way affect your academic progress. In return for your participation, you will be eligible to receive a $10 gift card to Tim Hortons.

This form outlines the procedures involved in this research, the risks and benefits associated with participation and what you can expect as a participant. Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator – researchethics@uoit.ca or 905.721.8668 x. 3693.

**Study Procedures:**

Participants are expected to arrive to the laboratory dressed in, or prepared to change into tight fitting shorts (spandex is preferred) and either no shirt (males only) or a tight fitting shirt that may be rolled up to expose the low back area. The Oswestry Low Back Disability Index (ODI) and demographic information forms will have already been completed and returned to the researchers prior to the study date. At the beginning of the study, participants will complete the Waterloo Handedness and Footedness questionnaires. Next, you will be fitted with instrumentation and your height and weight will be measured. The instruments that will be used during the study includes electromyography (EMG), motion capture (Optotrak) and force platforms. These instruments will measure your muscle activity, track your posture and measure your stability respectively.

Participants will complete maximum voluntary contractions (MVC's) for the gluteus medius and lumbar erector spinae to determine the peak muscle activity of these muscles. Next, participants will complete full low lumbar range of motion (ROM) tasks followed by a short seated rest break prior to the experimental trials.

When the study procedures begin, participants will be asked to stand at a desk while they complete a standardized computer task for 15 minutes without moving their feet. During the trial, participants will be asked to rate their low back discomfort and their hip angle will be measured by the investigator three times throughout the trial. The participants will then be randomized to complete the other three experimental trials where they will stand for 15 minutes completing a computer task while using various footrests. Each trial will be separated by 5 minute seated rest breaks. The footrest height will vary between trials. Once you have completed the four trials, you are free to go and will be contacted about the results of the research if you wish.

**Potential Benefits:**

If you decide to participate, you will be able to test out a standing desk without having to buy one. Sit-stand desks can be quite expensive, and this could be an excellent opportunity to see if it is worth buying one for yourself. Secondly, your participation will help improve the recommendations surrounding standing desks. Your participation will contribute towards the future direction of ergonomic recommendations and will help inform public opinion about the potential benefits of using a footrest while standing.

**Potential Risks:**

There are very few risks associated with participation in the study. The main task requires you to stand for 15 minutes. This task is rather common in everyday life and does not pose a great threat to your health. However, you should know that fatigue may set in during the study and if any severe discomfort occurs, notify the investigators immediately. Remember that you can withdraw from the study at any time, for any reason, without penalty. Your safety is the number one priority and you should not feel obligated to continue if you are in severe discomfort. However, you may experience mild discomfort with this task and will be asked to report your comfort levels periodically.

Secondly, the surface EMG markers pose a very low risk of skin irritation from the alcohol swab, razor, light abrasion, electrode gel or tape. These complications are not serious and they should subside within a few days. Participants will have access to soap and water to cleanse the affected area if this occurs.

However, if these irritations persist, we recommend that the participant goes directly to the campus health clinic for medical advice and then contact the researchers to report the adverse event.

**Confidentiality:**

Identifiers will be removed from all data to maintain confidentiality of the participants. The data will be stored in a locked are at UOIT for seven years from the completion of the study after which it will be destroyed in accordance with university protocol.

**Right to Withdraw:**

Your participation in this study is voluntary and you are free to decline without providing a reason. Throughout the research process, you are free to withdraw from participation at any time without repercussion.

**Debriefing and Dissemination of Results:**

The intent of this research is to improve guidelines. As such, the data for this research may be submitted to scientific conferences and peer reviewed journals for publication. Published data will be coded and no personal identifiers will be included. If you wish to receive an aggregate of the research findings, please check the box at the bottom of this form and provide an email address to receive the results.

**Thank You!**

Thank you very much for your time and help in making this study possible. If you have any questions concerning the research study, please contact the researcher Andrew Cregg at 647.285.4878 or Andrew.cregg@uoit.ca. Alternatively, you can contact the principal investigator Dr. Lori Livingston at 905.721.8668 or lori.livingston@uoit.ca.

This study has been approved by the UOIT Research Ethics Board REB [insert REB # assigned] on [insert date]. Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator –researchethics@uoit.ca or 905.721.8668 x. 3693.

**Sincerely,**

**Dr. Lori Livingston**
Dean, Faculty of Health Sciences
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4
Phone #: 905.721.8668
Email: lori.livingston@uoit.ca

**Andrew Cregg BSc, DC, MHSc (Candidate)**
Graduate Student, Faculty of Health Sciences
University of Ontario Institute of Technology (UOIT)
2000 Simcoe St. N., Oshawa, ON, L1H 7K4
Phone #: 647.285.4878
Email: Andrew.cregg@uoit.ca

***Please read the following carefully before signing. If you would like a copy of this consent form for your records, please ask the investigators.***

*Received Copy:* *YES* ☐ *NO* ☐

**I understand that:**

- Taking part in this study is voluntary and that I am free to withdraw from participation at any time without giving a reason and that withdrawal will in no way affect my academic process.
- This consent form will be kept in a locked area at UOIT, Oshawa, Ontario for seven years before being destroyed.
- Data collected during the study will be coded, kept in a confidential form and kept in a locked area at UOIT, Oshawa, Ontario for seven years before being destroyed.
- I may withdraw from participation at any time before, during or after the study up to two days following data collection. At which time my data will be included in the study.
- My participation in this study is confidential and that no material which could be used to identify me will be reported.

**I have:**

- Read and I understand the information provided within this consent form.
- Had the opportunity to ask questions and discuss the study with the investigators and am satisfied with the answers provided.
- Had time to consider whether or not to participate.
- Taken note of who to contact if I experience any adverse events.

**I give consent for the data from this study to be used in future research as long as there is no way that I can be identified in this research.** **YES** ☐ **NO** ☐

**I would like to receive a short report about the outcomes of this study. (If you answer yes, please provide an email)_____** **YES** ☐ **NO** ☐

By signing this form, you consent to participate in the study and you indicate that you understand the information provided to you within this document.

| _____ | _____ | _____ |
|---|---|---|
| Participants Name (Print) | Signature of Participant | Date |
| _____ | _____ | _____ |
| Witness' Name (Print) | Signature of Witness | Date |

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To be signed by the Primary Investigator and/or Student Lead:

I have fully explained the study to the participant to the best of my ability. I have provided ample opportunities for the participant to ask questions and I have provided clear answers. It is my opinion that the participant fully understands the requirements of the study, the potential risks and benefits of the study. The participant has provided voluntary consent and was not coerced into taking part in the study.

| _____ | _____ |
|---|---|
| Signature of the Investigator/Student Lead | Date |

## APPENDIX B: WATERLOO HANDEDNESS QUESTIONNAIRE

**Instructions**: Please indicate your hand preference for the following activities by circling the appropriate response. If you **always** (i.e. 95% or more of the time) use one hand to perform the described activity, circle **Ra** or **La** (for **right always** or **left always**). If you **usually** (i.e. about 75% of the time) use one hand circle **Ru** or **Lu** as appropriate. If you use both hands **equally often** (i.e. you use each hand about 50% of the time), circle **Eq**.

| | | | | | | |
|---|---|---|---|---|---|---|
| 1. | Which hand would you use to adjust the volume knob on a radio? | La | Lu | Eq | Ru | Ra |
| 2. | With which hand would you use a paintbrush to paint a wall? | La | Lu | Eq | Ru | Ra |
| 3. | With which hand would you use a spoon to eat soup? | La | Lu | Eq | Ru | Ra |
| 4. | Which hand would you use to point to something in the distance? | La | Lu | Eq | Ru | Ra |
| 5. | Which hand would you use to throw a dart? | La | Lu | Eq | Ru | Ra |
| 6. | With which hand would you use the eraser on the end of a pencil? | La | Lu | Eq | Ru | Ra |
| 7. | In which hand would you hold a walking stick? | La | Lu | Eq | Ru | Ra |
| 8. | With which hand would you use an iron to iron a shirt? | La | Lu | Eq | Ru | Ra |
| 9. | Which hand would you use to draw a picture? | La | Lu | Eq | Ru | Ra |
| 10. | In which hand would you hold a mug full of coffee? | La | Lu | Eq | Ru | Ra |
| 11. | Which hand would you use to hammer a nail? | La | Lu | Eq | Ru | Ra |
| 12. | With which hand would you use the remote control for a TV? | La | Lu | Eq | Ru | Ra |
| 13. | With which hand would you use a knife to cut bread? | La | Lu | Eq | Ru | Ra |
| 14. | With which hand would you use to turn the pages of a book? | La | Lu | Eq | Ru | Ra |
| 15. | With which hand would you use a pair of scissors to cut paper? | La | Lu | Eq | Ru | Ra |
| 16. | Which hand would you use to erase a blackboard? | La | Lu | Eq | Ru | Ra |
| 17. | With which hand would you use a pair of tweezers? | La | Lu | Eq | Ru | Ra |
| 18. | Which hand would you use to pick up a book? | La | Lu | Eq | Ru | Ra |
| 19. | Which hand would you use to carry a suitcase? | La | Lu | Eq | Ru | Ra |
| 20. | Which hand would you use to pour a cup of coffee? | La | Lu | Eq | Ru | Ra |
| 21. | With which hand would you use a computer mouse? | La | Lu | Eq | Ru | Ra |
| 22. | Which hand would you use to insert a plug into an outlet? | La | Lu | Eq | Ru | Ra |
| 23. | Which hand would you use to flip a coin? | La | Lu | Eq | Ru | Ra |
| 24. | With which hand would you use a toothbrush to brush your teeth? | La | Lu | Eq | Ru | Ra |
| 25. | Which hand would you use to throw a baseball? | La | Lu | Eq | Ru | Ra |
| 26. | Which hand would you use to turn a doorknob? | La | Lu | Eq | Ru | Ra |
| 27. | Which hand would you use for writing? | La | Lu | Eq | Ru | Ra |
| 28. | Which hand would you use to pick up a piece of paper? | La | Lu | Eq | Ru | Ra |
| 29. | Which hand would you use a hand saw? | La | Lu | Eq | Ru | Ra |
| 30. | Which hand would you use to stir a liquid with a spoon? | La | Lu | Eq | Ru | Ra |
| 31. | In which hand would you hold an open umbrella? | La | Lu | Eq | Ru | Ra |
| 32. | In which hand would you hold a needle while sewing? | La | Lu | Eq | Ru | Ra |
| 33. | Which hand would you use to strike a match? | La | Lu | Eq | Ru | Ra |
| 34. | Which hand would you use to turn on a light switch? | La | Lu | Eq | Ru | Ra |
| 35. | Which hand would you use to open a drawer? | La | Lu | Eq | Ru | Ra |
| 36. | Which hand would you use to press buttons on a calculator? | La | Lu | Eq | Ru | Ra |
| 37. | Is there any reason (i.e. injury) why you have changed your hand preference for any of the above activities? | **YES/NO** | (circle one) | | | |
| 38. | Have you been given special training or encouragement to use a particular hand for certain activities? | **YES/NO** | (circle one) | | | |
| 39. | If you have answered YES for either Questions 37 or 38, please explain: | | | | | |

## APPENDIX C: WATERLOO FOOTEDNESS QUESTIONNAIRE

**Instructions**: Answer each of the following questions as best you can. If you *always* use one foot to perform the described activity, circle **Ra** or **La** (for **right always** or **left always**). If you **usually** use one foot circle **Ru** or **Lu**, as appropriate. If you use **both** feet **equally often**, circle **Eq**.

Please do not simply circle one answer for all questions, but imagine yourself performing each activity in turn, and then mark the appropriate answer. If necessary, stop and pantomime the activity.

| | | | | | | |
|---|---|---|---|---|---|---|
| 1. | Which foot would you use to kick a stationary ball at a target straight in front of you? | La | Lu | Eq | Ru | Ra |
| 2. | If you had to stand on one foot, which foot would it be? | La | Lu | Eq | Ru | Ra |
| 3. | Which foot would you use to smooth sand at the beach? | La | Lu | Eq | Ru | Ra |
| 4. | If you had to step up onto a chair, which foot would you place on the chair first? | La | Lu | Eq | Ru | Ra |
| 5. | Which foot would you use to stomp on a fast-moving bug? | La | Lu | Eq | Ru | Ra |
| 6. | If you were to balance on one foot on a railway track, which foot would you use? | La | Lu | Eq | Ru | Ra |
| 7. | If you wanted to pick up a marble with your toes, which foot would you use? | La | Lu | Eq | Ru | Ra |
| 8. | If you had to hop on one foot, which foot would you use? | La | Lu | Eq | Ru | Ra |
| 9. | Which foot would you use to help push a shovel into the ground? | La | Lu | Eq | Ru | Ra |
| 10. | During relaxed standing, people initially put most of their weight on one foot, leaving the other leg slightly bent. Which foot do you put most of your weight on first? | La | Lu | Eq | Ru | Ra |
| 11. | Is there any reason (i.e. injury) why you have changed your foot preference for any of the above activities? | **YES** | **NO** | (circle one) | | |
| 12. | Have you ever been given special training or encouragement to use a particular foot for certain activities? | **YES** | **NO** | (circle one) | | |
| 13. | If you have answered **YES** for either question 11 or 12, please explain: | | | | | |

# APPENDIX D: OSWESTRY LOW BACK DISABILITY INDEX (OBDI)

## OSWESTRY BACK DISABILITY INDEX

PATIENT NAME: _____ FILE #: _____ DATE: _____

Please rate the severity of your back pain by circling a number below.

| 0 | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 |
|---|---|---|---|---|---|---|---|---|---|---|

No pain — Unbearable pain

**Instructions: Please mark the ONE BOX in each section, which most closely describes your problem.**

### Section 1 – Pain Intensity

- ☐ 0. The pain comes and goes and is very mild.
- ☐ 1. The pain is mild and does not vary much.
- ☐ 2. The pain comes and goes and is moderate.
- ☐ 3. The pain is moderate and does not vary much.
- ☐ 4. The pain comes and goes and is severe.
- ☐ 5. The pain is severe and does not vary much.

### Section 2 – Personal Care (Washing, Dressing, etc.)

- ☐ 0. I would not have to change my way of washing or dressing in order to avoid pain.
- ☐ 1. I do not normally change my way of washing or dressing even though it causes some pain.
- ☐ 2. Washing and dressing increase the pain but I manage not to change my way of doing it.
- ☐ 3. Washing and dressing increase the pain and I find it necessary to change my way of doing it.
- ☐ 4. Because of the pain, I am unable to do some washing and dressing without help.
- ☐ 5. Because of the pain, I am unable to do any washing or dressing without help.

### Section 3 – Lifting

- ☐ 0. I can lift heavy weights without extra pain.
- ☐ 1. I can lift heavy weights but it gives extra pain.
- ☐ 2. Pain prevents me from lifting heavy weights off the floor.
- ☐ 3. Pain prevents me lifting heavy weights off the floor, but I can manage if they are conveniently positioned, e.g., on a table.
- ☐ 4. Pain prevents me from lifting heavy weights, but I can manage light to medium weights if they are conveniently positioned, e.g., on a table.
- ☐ 5. I can lift only very light weights.

### Section 4 – Walking

- ☐ 0. I have no pain when walking.
- ☐ 1. I have some pain when walking, but it does not increase with distance.
- ☐ 2. I cannot walk more than 1 mile without increasing pain.
- ☐ 3. I cannot walk more than ½ mile without increasing pain.
- ☐ 4. I cannot walk more than ¼ mile without increasing pain.
- ☐ 5. I cannot walk at all without increasing pain.

### Section 5 – Sitting

- ☐ 0. I can sit in any chair for as long as I like.
- ☐ 1. I can sit only in my favourite chair for as long as I like.
- ☐ 2. Pain prevents me from sitting for more than 1 hour.
- ☐ 3. Pain prevents me from sitting for more than 1/3 hour.
- ☐ 4. Pain prevents me from sitting more than 10 minutes.
- ☐ 5. I avoid sitting because it increases pain immediately.

### Section 6 – Standing

- ☐ 0. I can stand for as long as I want without pain.
- ☐ 1. I have some pain on standing but it does not increase with time.
- ☐ 2. I cannot stand longer than 1 hour without increasing pain.
- ☐ 3. I cannot stand longer than ½ hour without increasing pain.
- ☐ 4. I cannot stand longer than 10 minutes without increasing pain.
- ☐ 5. I avoid standing because it increases the pain immediately.

### Section 7 – Sleeping

- ☐ 0. I have no pain in bed.
- ☐ 1. I have pain in bed, but it does not prevent me from sleeping well.
- ☐ 2. Because of my pain, my normal night's sleep is reduced by less than ¼.
- ☐ 3. Because of my pain, my normal night's sleep is reduced by less than ½.
- ☐ 4. Because of pain, my normal night's sleep is reduced by less than ¾.
- ☐ 5. Pain prevents me from sleeping at all.

### Section 8 – Social Life

- ☐ 0. My social life is normal and gives me no pain.
- ☐ 1. My social life is normal but increases the degree of pain.
- ☐ 2. Pain has no significant effect on my social life apart from limiting my more energetic interests e.g., dancing, etc.
- ☐ 3. Pain has restricted my social life and I do not go out often.
- ☐ 4. Pain has restricted my social life to my home.
- ☐ 5. I have hardly any social life because of the pain.

### Section 9 – Travelling

- ☐ 0. I have no pain when travelling.
- ☐ 1. I have some pain when travelling but none of my usual forms of travel make it any worse.
- ☐ 2. I have extra pain when travelling but it does not compel me to seek alternative forms of travel.
- ☐ 3. I have extra pain when travelling which compels me to seek alternative forms of travel.
- ☐ 4. Pain restricts me to short necessary journeys of fewer than 30 minutes.
- ☐ 5. Pain restricts all forms of travel.

### Section 10 – Changing Degree of Pain

- ☐ 0. My pain is rapidly getting better.
- ☐ 1. My pain fluctuates but overall is definitely getting better.
- ☐ 2. My pain seems to be getting better but improvement is slow.
- ☐ 3. My pain is neither getting better nor worse.
- ☐ 4. My pain is gradually worsening.
- ☐ 5. My pain is rapidly worsening.

*Revised: Aug 2002*

Clinician Review _____ Score ☐

## APPENDIX E: DEMOGRAPHIC INFORMATION

***Title: Changes in lumbar spine posture, muscle activation and centre of pressure while standing with different footrest heights during a standardized computer task.*** *This study has been approved by the UOIT Research Ethics Board REB [#14477] on August 14th, 2017.*

***If you would like a copy of this consent form for your records, please ask the investigators.***

***Received Copy:*** ***YES*** ☐ ***NO*** ☐

**Name: _____ Gender (Circle one): Male Female**

**Date of Birth: _____ Age: _____ Height: _____**

**Email Address: _____**

| | | |
|---|---|---|
| Have you experienced low back pain in the last 12 months? | **YES** ☐ | **NO** ☐ |
| If you answered yes, has that back pain caused you to seek medical treatment or take more than 3 days off work? | **YES** ☐ | **NO** ☐ |
| Have you worked a job over the last 12 months that requires you to stand still for longer than 2 hours? | **YES** ☐ | **NO** ☐ |
| Have you ever had low back or hip surgery? | **YES** ☐ | **NO** ☐ |
| Are you able to stand for two hours without significant discomfort? | **YES** ☐ | **NO** ☐ |
| Would you like to be notified with the aggregate results of the study when they are released in early 2018 via email? | **YES** ☐ | **NO** ☐ |
| I hereby give consent for the information contained in this package to be used for the purposes of this study and in future research as long as there is no way that I can be identified**.** | **YES** ☐ | **NO** ☐ |

If you have any questions concerning the research study, please contact the researcher Andrew Cregg at 647.285.4878 or Andrew.cregg@uoit.ca. Alternatively, you can contact the principal investigator Dr. Lori Livingston at 905.721.8668 or lori.livingston@uoit.ca.

Any questions regarding your rights as a participant, complaints or adverse events may be addressed to Research Ethics Board through the Research Ethics Coordinator –researchethics@uoit.ca or 905.721.8668 x. 3693.

_____ _____

**Participant Signature** **Date**

## APPENDIX F: DISCOMOFRT QUESTIONNAIRE (VAS & RPD)

Please place a mark on the line to indicate the CURRENT level of your pain in your LOW BACK

No Pain — Worst Pain Imaginable