THE EFFECTS OF INSOLES ON TESTS OF LUMBOPELVIC CONTROL AND BALANCE, AND THE ASSOCIATION BETWEEN THESE TESTS

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Abstract

The first purpose of this thesis was to examine the effects of insoles on lumbopelvic control using biomechanical and test performance measures during the sidelying active hip abduction test, and two standing balance tests (the one-leg stance test and the Sharpened Romberg test). The use of a relative change analysis allowed for the targeted examination of somatosensory mechanisms thought to underlie improved lumbopelvic control with insole use. The second was to examine the relationship between the sidelying test and standing balance tests. Perceived low back pain was examined in association with each of these purposes. In general, the use of insoles in the unaffected university aged participants produced minimal and mixed results for all of the measures examined, with no change in perceived low back pain. No association between the sidelying test and balance tests was observed, as well as no association between these tests and perceived low back pain.

Dedication

I would like to dedicate my Master's thesis to my family: my dad George, my mom Annette, my sister Mary-Anne, my brother-in-law Jeremy, and my brother Andrew for their constant love and support in life and throughout my entire academic career, especially during my MSc. Thank you for so many things. Thanks for being patient with me during moments of frustration, for building me up when I was low, for indulging me during my moments of absolute nerdiness (with much appreciated nods, smiles, and mhmmms), for the extraordinary amount of enthusiasm you always expressed for my achievements and finally, for always believing in me even when I did not believe in myself. Mom, I will be forever thankful for every single phone call and each and every moment that you got me through many very trying days. There are no words which can explain how grateful I truly am.

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My apologies if I have missed anyone!

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List of Abbreviations

Abbreviation	**Term**
AHAbd	Active hip abduction
ANOVA	Analysis of Variance
A-P	Antero-posterior
ASLR	Active Straight Leg Raise
BF	Biceps femoris
BMI	Body Mass Index
BOS	Base of support
CNS	Central nervous system
COP	Centre of pressure
COP_d	Centre of pressure displacement
COP_{RMS}	Centre of pressure root mean square
$COP_{totalpath}$	Centre of pressure total path
COP_v	Centre of pressure velocity
COM	Centre of mass
EMG	Electromyography
EO	External oblique
ES	Erector spinae
GTMED	Gluteus medius
ICC	Intraclass correlation coefficient
IO	Internal oblique
LES	Lumbar erector spinae
LBP	Low back pain
MCID	Minimally clinically important difference
M-L	Medio-lateral
MVC	Maximum voluntary contraction
NPD	Non-pain developers
OLST	One-leg stance test
PD	Pain developers

RA	Rectus abdominis
R/L	Right and left
SD	Standard deviation
SEM	Standard error of measurement
SRT	Sharpened Romberg test
TES	Thoracic erector spinae
VAS	Visual analogue scale
VM	Vastus medialis

CHAPTER 1

Global Introduction

1. Global Introduction

Low back pain (LBP) is one of the leading musculoskeletal disorders, with a lifetime prevalence of up to 84% (Airaksinen et al., 2006). Globally, it causes more disability than any other condition (Hoy et al., 2014), and it imposes a tremendous economic burden of up to an estimated $100 billion for direct (medical) and indirect (i.e. lost wages and reduced productivity) costs annually in the United States alone (Frymoyer & Cats-Baril, 1991). To try and ameliorate the serious human and economic costs of LBP, it is critical to understand its causal and compensatory mechanisms to improve the prevention and treatment of it. While about 90% of all LBP is due to an unknown cause (Koes, van Tulder, & Thomas, 2006), biomechanical abnormalities have been suggested to account for this LBP (Kendall, Schmidt, & Ferber, 2014). The identification of clinical tests and their associated biomechanical measures, which may be predictive of LBP development, can have valuable implications for clinical prevention strategies. Beyond serving as clinical markers for LBP development, improvements in the biomechanical performance of these tests as a result of a targeted intervention provides support for the use of these tests and their associated measures in LBP prevention, and support for interventions which are effective in improving these tests and measures.

Insoles have been commonly prescribed to treat several conditions including pain in the foot, knee, hip, and back (Walter, Ng, & Stoltz, 2004). While the literature is mixed, insoles have been shown to prevent (Larsen, Weidich, & Leboeuf-Yde, 2002) and reduce (Cambron, Duarte, Dexheimer, & Solecki, 2011; Shabat, Gefen, Nyska, Folman, & Gepstein, 2005) LBP. Most of this evidence is based on self-reported LBP, with no empirical biomechanical evidence to support functional changes which may contribute to these findings. When biomechanical measures such as surface electromyography (EMG) (Murley, Buldt, Trump, & Wickham, 2009;

Tomaro & Burdett, 1993) and kinematics (McClay & Manal, 1998) have been examined, there has been a heavy focus on the lower extremities. Despite the association between lumbopelvic control and LBP (Luomajoki & Moseley, 2011; Nelson-Wong, Flynn, & Callaghan, 2009), and the important role this region plays in postural control (Thorstensson, Carlson, Zomlefer, & Nilsson, 1982; Winter, 1995a), there remains a dearth in the literature regarding the effects of insoles on the lumbopelvic region. For the little evidence that exists, the effects of insoles on lumbopelvic EMG outcomes are mixed (Bird, Bendrups, & Payne, 2003; Hertel, Sloss, & Earl, 2005), and on lumbopelvic kinematic outcomes are minimal (Marinakis & Catalfamo, 2004; Nester, van der Linden, & Bowker, 2003) and absent (Esfandiari, Kamyab, Tazdi, Forough, & Sanjari, 2013).

Most biomechanical insole studies have focused on gait (Murley & Bird, 2006; Tomaro & Burdett, 1993), with far fewer studies assessing the effects of these devices on postural control during standing balance tests. One of the major mechanisms these devices are designed to treat involves the suggested link between abnormal foot mechanics and LBP based on the kinematic chain (Bird & Payne, 1999). However, this link is heavily theoretical and as a result, the biomechanical effects of insoles are not well understood. Another mechanism is based on studies that have assessed the effect that insoles have on balance related outcomes. In these studies it is suggested that insoles enhance the somatosensory (tactile and proprioceptive) information from the plantar surface of the foot (Hamlyn, Docherty, & Klossner, 2012; Hijmans, Geertzen, Dijkstra, & Posterma, 2007). It is the enhanced somatosensory input that is thought to inform changes in neuromuscular control that lead to improvements in balance and posture during standing (Hamlyn et al., 2012; Hijmans et al., 2007; Mochizuki & Amadio, 2003; Olmstead & Hertel, 2004; Rothbart, 2005). Although the few studies which have examined balance outcomes have produced mixed results, many have observed improvements in balance through reductions in centre of pressure (COP) measures (Hamlyn et al., 2012; Losa Iglesias, Becerro de Bengoa Vallejo, & Palacios Peña, 2012; Qiu et al., 2013). Given the lacking and inconclusive literature investigating the effects of insoles on the lumbopelvic region and on standing balance performance, coupled with the potential of these devices to improve LBP outcomes, this area of research requires further investigation. As a means of targeting the hypothesis that changes in somatosensory information can drive improvements in neuromuscular control, the first aim of the current study included an examination of the influence that insoles have on balance under conditions in which the sensory information has been reweighted (towards proprioception). This was achieved by challenging the somatosensory system using an eyes closed condition in which no visual information was available, and also using proprioceptively challenging balance tests with a narrow and/or small base of support (BOS).

Lumbopelvic control is a critical component of balance in many movements and postures of daily living (Sahrmann, 2002; Thorstensson et al., 1982). While tests of lumbopelvic control and standing balance have been assessed separately within a LBP and low back injury context (Olivier, Stewart, Olorunju, & McKinon, 2015), to the author’s knowledge, no studies to date have examined the association between lumbopelvic control during an active movement performed in a sidelying position and balance during upright standing. The active hip abduction (AHAbd) test, which assesses lumbopelvic control in a sidelying position, is a novel screening tool designed to identify occupational LBP during prolonged standing (Nelson-Wong et al., 2009). Within a LBP paradigm, this test has proven to be sensitive to impairments in lumbopelvic control as it was the only test out of a wide array of clinical and performance-based measures to show an association with LBP by significantly distinguishing between groups

(Babiolakis, Kuk, & Drake, 2015; Nelson-Wong et al., 2009). Worse lumbopelvic control (higher AHAbd test scores) was observed in previously asymptomatic (back-healthy) individuals who developed a clinically relevant level of transient LBP during prolonged standing compared to those who did not (Nelson-Wong et al., 2009) and in nurses with a recent history of LBP due to back injury compared to those without this LBP history (Babiolakis et al., 2015).

Based on the heavy reliance of specific standing balance tests such as the one-leg stance test (OLST) and Sharpened Romberg test (SRT) on lumbopelvic control as a result of the postures required (OLST: small and narrow BOS, SRT: narrow BOS) and for the OLST in particular, with the shift from double to single stance (Lee, 2011; Winter, 1995a), impairments in lumbopelvic control in a sidelying position may translate to impairments in lumbopelvic control in upright standing, functional positions. It was therefore, a second aim of this study to assess whether there was an association between the sidelying hip abduction test and standing balance tests that might be more functionally relevant as lumbopelvic assessment tools for populations in which prolonged standing induces LBP.

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1.1 Research Questions and Hypotheses

1.1.1 Research Questions

There were two main purposes of this thesis, both centred on lumbopelvic control. The first was to quantify and assess the effects of insoles on lumbopelvic control using biomechanical (muscle activation, three-dimensional kinematic, kinetic), and test performance outcomes during three clinical tests, a sidelying test of lumbopelvic control, and two standing balance tests. The second was to quantify the relationship between the sidelying test and two standing balance tests. In relation to these two main purposes, it was also of interest to examine the effects of the insoles on perceived LBP during standing, and to determine the association between the three clinical tests and perceived LBP. Based on these purposes, the following research questions were addressed using two data collections (1) and one data collection (2):

1. How do insoles affect the following measures after eight weeks of use:

   A. Lumbopelvic and thigh muscle activations during the sidelying AHAbd test, and two standing balance tests, the OLST, and SRT?

   B. Three-dimensional lumbopelvic kinematics during the OLST?

   C. Kinetic measures during the OLST and SRT?

   D. Test performance outcomes for the AHAbd test, OLST, and SRT?

   E. Perceived LBP during standing?

2. Is performance on the AHAbd test related to performance on the OLST and SRT, and are all three of these clinical tests related to perceived LBP during standing?

1.1.1 Hypotheses

Before and after eight weeks of insole use, a comprehensive set of measures were assessed during three clinical tests. Trunk, hip, and thigh muscle activations were examined for the AHAbd test, OLST, and SRT. Trunk and pelvis three-dimensional kinematics (in each axis separately, and combined) were assessed for the OLST. Kinetic COP measures (displacement, root mean square, total path, and velocity) were analyzed for both the OLST and SRT. Performance outcomes (rated scores/times/pelvic drop) were assessed for all three clinical tests. Finally, perceived LBP (visual analogue scale (VAS) scores) during standing was also examined. With these measures, the following hypotheses were developed in response to the above general research questions:

1. Following eight weeks of use, insoles will:

A. Decrease the maximum surface EMG activity of the trunk, hip, and thigh muscles during the AHAbd test, and will improve balance (likely due to proprioceptive mechanisms) during the OLST and SRT as reflected by a decrease in the difference (relative change) between the eyes closed and eyes open conditions for the maximum and mean surface EMG activity of these same muscles.

B. Improve balance (likely due to proprioceptive mechanisms) as reflected by a decrease in the difference (relative change) between the eyes open and eyes closed conditions for maximum and mean three-dimensional lumbar, pelvis, lumbopelvic, and trunk angles (in each axis separately, and combined as a Euclidean norm angle) during the OLST.

C. Improve balance (likely due to proprioceptive mechanisms) as reflected by a decrease in the difference (relative change) between the eyes open and eyes closed conditions for

COP (displacement, root mean square, velocity, and total path) during the OLST and SRT.

D. Improve performance on the AHAbd test as reflected through a decrease in the score and will improve performance on the OLST and SRT (likely due to proprioceptive mechanisms) as reflected by a decrease in the difference (relative change) between the eyes open and eyes closed conditions for the balance times and a decrease in the presence of pelvic drop during the OLST.

E. Decrease perceived LBP (VAS scores) during standing.

2. Performance on the AHAbd test (pass/fail based on scores) will be related to performance on the OLST and SRT (pass/fail based on balance times) during visit one, and performance on all three of these clinical tests (scores/balance times) will be correlated to maximum perceived LBP (VAS scores) during standing in visit one.

CHAPTER 2

Review of Related Literature

2. Review of Related Literature

The following chapter begins with a review of the literature required to understand the methodology and clinical applications of a functionally induced transient LBP model. Following this, there will be an overview of relevant lumbopelvic anatomy, the major causes and consequences of general functional impairments, and the specific discussion of these impairments as they pertain to lumbopelvic control and balance and foot function within a LBP paradigm. The emphasis within this section will be on the discussion of particular lumbopelvic and balance tests as, well as insoles interventions to give context for the use of these tests and this intervention within the present study.

2.1 Functionally Induced Transient Low Back Pain Model

The theory, methodology, and clinical applications of a transient LBP model will be discussed to provide the context of its use within the present study.

2.1.1 Transient Low Back Pain Protocol

Many studies investigate LBP by comparing individuals with LBP to back-healthy controls (Braga et al., 2012; Brumagne, Cordo, Lysens, Verschueren, & Swinnen, 2000; Scholtes, Gombatto, & Van Dillen, 2009). However, with these types of cross-sectional study designs, it is not possible to discern whether impairments in those with LBP are causal or compensatory. Therefore, a study design which is prospective to infer causation, yet short in duration, can be powerful in terms of its implications. Using this approach combined with the knowledge that prolonged standing has a strong association with LBP and low back discomfort (Macfarlane et al., 1997; Ryan, 1989), Gregory, Brown, and Callaghan (2008) developed a two hour prolonged standing protocol to functionally induce transient low back discomfort in previously asymptomatic (back-healthy) individuals. To be specific, these asymptomatic (back-healthy) individuals had no history of low back injury or LBP over the previous 12 months and were free of any low back discomfort at the start of the study (Gregory et al., 2008). Recent work using this model more explicitly described the LBP history of these individuals as having no lifetime event of LBP that was significant enough to seek medical care or that resulted in greater than three days off from work or school; no current low back or hip pain, and no previous hip surgery (Nelson-Wong & Callaghan 2010abc; Nelson-Wong et al., 2009; Nelson-Wong, Howarth, & Callaghan, 2010). The present study used these same inclusion criteria for participants. With only a percentage of the sample developing transient LBP during standing (Nelson-Wong & Callaghan, 2010b), researchers can make comparisons between those who do (PDs: pain developers) and do not (NPDs: non-pain developers) develop a clinically relevant level of transient LBP. As a result, in just a few hours, researchers are able to investigate whether specific alterations or impairments in PDs were present before (causal) or after (compensatory) their development of the pain. This is in contrast with typical prospective designs, where it would take months or years to investigate this. The AHAbd test will be discussed as a causal factor in relation to this transient LBP model in Section 2.3.2.

This protocol involves a prolonged standing occupational simulation. While standing on a confined work surface (i.e. 0.50m x 0.46m) for two hours, asymptomatic (back-healthy) participants perform light simulated work tasks which are reflective of jobs that often require periods of prolonged standing – i.e. small object assembly to mimic an assembly line worker, currency sorting to mimic a bank teller, and card dealing to mimic a casino dealer (Gallagher & Callaghan 2015; Gallagher et al., 2011; Gregory et al., 2008; Gregory & Callaghan, 2008; Marshall, Patel, & Callaghan, 2011; Nelson-Wong & Callaghan, 2010abcd; Nelson-Wong, Alex, Csepe, Lancaster, & Callaghan, 2012b; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008;

Nelson-Wong et al., 2010; Raftry & Marshall, 2012; Sorensen, Johnson, Callaghan, George, and Van Dillen, 2015). Participants are instructed to stand in their usual, comfortable manner, however, they are not permitted to rest their arms or legs on the work table in front of them. In the original protocol (Gregory et al., 2008), participants indicated their level of perceived low back discomfort prior to and following the standing period. Subsequent studies assessed participants' perceived pain more frequently, with measurements at baseline and every 15 minutes throughout the protocol (Marshall et al., 2011; Nelson-Wong & Callaghan, 2010abcd; Nelson-Wong et al., 2012b; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008; Nelson-Wong et al., 2010; Raftry & Marshall, 2012; Sorensen et al., 2015). The present study used a similar protocol, with participants completing a card dealing task, receiving the same instructions regarding their standing posture, and marking their perceived LBP at baseline and every 15 minutes during standing. However, as prolonged standing was not the major focus of this study, the duration of the standing stimulus was markedly shorter than previous work, with participants only standing for 30 minutes in total. A moderate to high percentage (40-71%) of PDs were observed in previous studies which used the same VAS PD/NPD classification as the present study (Gallagher et al., 2011; Marshall et al., 2011; Nelson-Wong & Callaghan, 2010abc; Nelson-Wong et al., 2009; Raftry & Marshall, 2012). The quantification of PDs and NPDs using the VAS will be discussed further below.

2.1.2 Quantification of Transient Low Back Pain and Clinical Applications

Studies which used this transient LBP model quantified low back discomfort or pain using the VAS. It is a valid (Summers, 2001) and reliable (Revill, Robinson, Roden, & Hogg, 1976) measure which is easy to use and has been used extensively in research (Gallagher et al., 2011; Marshall et al., 2011; Nairn, Azar, & Drake, 2013; Nelson-Wong & Callaghan, 2010abc;

Nelson-Wong et al., 2009; Nelson-Wong et al., 2008; Raftry & Marshall, 2012; Schinkel-Ivy, Nairn, & Drake, 2013) and clinical settings (Mohan, Ryan, Whelan, & Wakai, 2010; Scrimshaw & Maher, 2001). As the terms 'pain' and 'discomfort' have both been used for the VAS, it is important to distinguish the use and association of the these terms. The term 'discomfort' has been used for the VAS in earlier studies (Gregory et al., 2008; Nelson-Wong et al., 2008), however, the term 'pain' was used in the present study in line with the more recent literature (Nelson-Wong et al., 2009; Nelson-Wong et al., 2010; Gallagher, Nelson-Wong, & Callaghan, 2011; Raftry & Marshall, 2012). While the use of these terms can be controversial in the literature, they have been treated synonymously in clinical studies (Tait & Chibnall, 2002; Schmader et al., 2007) and discomfort assessment tools have been shown to be highly and significantly correlated with pain assessment tools (Crane et al., 2005).

To classify previously asymptomatic (back-healthy) individuals as PDs or NPDs using the VAS, a clinically relevant level (cut-point) of transient LBP was determined. This cut-point was originally based on clinical work which reported that the minimally clinically important difference (MCID) for patients to perceive a worsening of their pain in response to treatment was 8 mm (Hagg, Fritzell, & Nordwall, 2003) and 9 mm (Kelly, 1998) on a 100 mm VAS. According to Kelly (1998), it is unlikely that reported changes in perceived pain less than this amount are of clinical importance. To be conservative, studies using the transient LBP protocol classified their PDs and NPDs based on a larger MCIDs (higher cut-point) of 10 mm (Gallagher et al., 2011; Marshall et al., 2011; Nelson-Wong & Callaghan, 2010abc; Nelson-Wong et al., 2009; Raftry & Marshall, 2012).Given the clinically determined MCIDs, and the application of MCIDs in previous transient LBP models studies, a change of 10 mm from baseline on the VAS was used as a conservative cut-point to distinguish between PDs (>10 mm) and NPDs (≤10 mm) in the present study.

While this transient LBP model is valuable in assessing potential causal factors for LBP in a short period of time, the LBP observed is experimentally induced, transient (it goes away once the standing task is terminated), and subjective (Gallagher & Callaghan 2015). Thus, an important consideration is whether this model has clinical applications. Using a three year longitudinal study, Nelson-Wong and Callaghan (2014) determined that the transient LBP model is predictive of future clinical LBP in previously asymptomatic (back-healthy) individuals. Compared to NPDs, PDs reported significantly higher rates of clinical LBP (35.3% vs. 23.1%) and based on a diagnostic odds ratio, were three times more likely to experience an episode of clinical LBP during the first two years of follow-up (Nelson-Wong & Callaghan, 2014). Based on self-reported questionnaire data from this study, none of the clinical LBP was a result of trauma (i.e. a motor vehicle accident) and likely had an insidious onset (Nelson-Wong & Callaghan, 2014). Another important consideration is the validity of this transient LBP model. To examine this, researchers determined whether during the transient LBP model, people with clinical LBP reported symptoms similar to (1) their ‘typical’ symptoms, and to (2) symptoms reported by previous asymptomatic (back-healthy) individuals who became PDs with this same model (Sorensen et al., 2015). Sorensen and colleagues (2015) concluded that the symptoms experienced by people with clinical LBP during this model were similar to both their ‘typical’ symptoms and to the symptoms reported by PDs, providing evidence for the validity of this paradigm.

2.1.3 Functionally Induced Low Back Pain Model Summary

Many studies which investigate LBP are cross-sectional and cannot infer causation. Using a prolonged standing occupational simulation, the transient LBP model is a powerful paradigm which is prospective to infer causation, yet short in duration. With only a percentage of previously asymptomatic (back-healthy) individuals developing LBP (PDs) during this protocol, researchers are able to make comparisons between those who do (PDs) and do not (NPDs) develop a clinically relevant (VAS determined) level of LBP, and can assess whether impairments in PDs were present before (causal) or after (compensatory) their development of the pain. While the LBP in this protocol is experimentally induced, transient in nature, and subjective, its PD/NPD cut-point is based on a clinical level of LBP, it is comparable to typical symptoms associated with clinical LBP, and it is predictive of future clinical LBP development.

2.2 Anatomy of the Trunk, Hip, and Thigh

When discussing the biomechanics of the lumbopelvic region, which consists of the musculoskeletal structures of the lumbar spine, pelvis, and hip, it is critical to understand its structural and functional anatomy. Within each sub-section, the bony and muscular anatomy of each region of interest (trunk/hip/thigh) will be reviewed.

2.2.1 Trunk

While the human vertebral column (spine) consists of five regions (cervical, thoracic, lumbar, sacrum, and coccyx), the main focus will be on the lumbar region. The normal curves within the spine are dynamic (changing shape during movements and with adjustments of posture) and have several roles (Neumann, 2010a). These curves define the neutral position of each spinal region, increase the strength and protection of the spine, help maintain balance in the upright position, and contribute to shock absorption during gait (Neumann, 2010a; Tortora & Nielsen, 2012). The neutral position of the lumbar spine in particular is convex (bulging out) anteriorly, and concave (cupping in) posteriorly, exhibiting an alignment referred to as lordosis, meaning ‘bent backward’ (Neumann, 2010a; Tortora & Nielsen, 2012). If this lumbar lordosis is excessive (i.e. accentuated by anterior pelvic tilt), it can lead to potentially injurious changes within the lumbar spine including LBP (Neumann, 2010a; Shirazi-Adl & Drouin, 1987).

In terms of its make-up, the lumbar spine consists of five vertebrae (L_1-L_5) which are the largest and strongest of the unfused vertebrae in the spine, and are designed to support the entire weight of the head, arms, and trunk (Neumann, 2010a; Tortora & Nielsen, 2012). Functionally, spine movements are described as rotations within three cardinal planes, about three axes of rotation, with flexion and extension, lateral flexion, and axial twist occurring in the sagittal plane (M-L: medio-lateral axis of rotation), frontal plane (A-P: antero-posterior axis of rotation), and transverse plane (supero-inferior axis of rotation), respectively (Neumann, 2010a). While the structure of the lumbar spine allows for a large range of motion in the sagittal plane (~ 55-70º of flexion and extension combined), it allows for substantially less motion in the frontal plane (~ 20º of lateral flexion each side), and very minimal motion in the transverse plane (i.e. ~ 5-7º of axial twist each side) (Neumann, 2010a; Troke, Moore, Maillardet, & Cheek, 2005).

Without the stabilizing function of its associated muscles, the osteoligamentous spine is incredibly unstable and cannot tolerate compressive load (Lucas & Bresler, 1961; McGill, 2007; Panjabi, 1992). Thus, the neuromuscular control of the lumbar spine is critical in its optimal functioning, and in stabilizing the lumbar vertebrae against failure (McGill, 2007). Muscles of the trunk (abdomen and back) have several important functions, playing a crucial role in spinal stability (see Section 2.3.1), and the biomechanical interactions between the spine, pelvis, and lower extremities, which are all critical components of lumbopelvic and postural control (Gardner-Morse & Stokes, 1998; Granata & Marras, 2000; McGill, 2007; Sahrmann, 2002).

The abdominal wall covers a relatively large area, spanning from the xiphoid process of the sternum and costal margins of the rib cage (superiorly), to the spine (posteriorly) and pelvis (inferiorly) (Drake et al., 2012). It is made up of five muscles which are positioned anteriorly and laterally on the trunk, including the rectus abdominis (RA), and the external and internal obliques (EOs and IOs, respectively) which are the focus of this review, along with the transversus abdominis, and the pyramidalis muscles (Drake et al., 2012). The rectus sheath is composed of a blending of connective tissues formed by the EO, IO, and transversus abdominis muscles, which surround the RA muscle, thickening and crisscrossing as they span across the midline to form the linea alba (Neumann, 2010a) This is an important structural consideration which strengthens the abdominal wall, and functionally links the right and left abdominal wall muscles, providing an effective way to transfer muscular force across the midline of the body (McGill, 2007; Neumann, 2010a; Sahrmann, 2002). While the abdominal wall muscles work together, they also can work independently, with individuals who exhibit substantial motor control in this region being able to differentially activate specific portions of the abdominal musculature (McGill, 2007; Stokes, Gardner-Morse, & Henry, 2011; Vera-Garcia, Moreside, & McGill, 2011). For example, some trained individuals are able to target the activation of the transversus abdominis alone, while relaxing the rest of the abdominal wall muscles which causes the drawing in and hollowing of the abdominal wall (McGill, 2007). In terms of its function, the RA is considered a major trunk flexor which also produces posterior pelvic tilt (Norris, 1993; Sahrmann, 2002). The major function of the obliques (EO, IO) involves trunk rotation (contralateral and ipsilateral, respectively) and trunk lateral flexion when activated unilaterally, as well as trunk flexion when activated bilaterally (Drake et al., 2012; McGill, 2007; Ng et al., 2001). Together, the abdominal wall muscles serve many important physiological functions, supporting and protecting the abdominal viscera, and compressing the abdomen to assist with active expiration, defecation, urination, and childbirth through an increase in intra-thoracic and intra-abdominal pressure (Tortora and Nielsen, 2012). The abdominal muscles also play a major role in the stabilization of the spine in all three planes of motion, ideally optimizing postural alignment, establishing a firm base for muscles to move the limbs, and limiting excessive, potentially stressful, and compensatory motion in the spine and extremities (Gardner-Morse & Stokes, 1998; Neumann, 2010a; Sahrmann, 2002). If even one of these trunk muscles inappropriately activates (magnitude and/or timing), the trunk can become unstable, leading to motor control impairments which have been associated with LBP (McGill, Grenier, Kavcic, & Cholewicki, 2003). For instance, if the RA becomes dominant, and the obliques are compromised, control of the trunk or pelvic rotation becomes impaired, as the RA cannot control the rotation (Drake et al., 2012; Sahrmann, 2002).

The erector spinae (ES) muscles are the largest posterior trunk muscles which span the trunk vertically (Tortora & Nielsen, 2012). From lateral to medial, the ES muscles consist of the iliocostalis, longissiumus, and the spinalis (Drake et al., 2012). These muscles have multiple origin and insertion sites at different levels of the rib cage, vertebrae (transverse and spinous processes), and pelvis, with most of them having a common attachment on a broad and thick common tendon located on the sacrum (Neumann, 2010a; Tortora & Nielsen, 2012). The primary role of the ES muscles is trunk extension when activated bilaterally, but they are also important in controlling trunk lateral flexion and axial twist when activated unilaterally (Tortora & Nielsen, 2012). Further, the ES muscles also cause anterior and lateral pelvic tilt through bilateral and unilateral activations, respectively (McGill, 2007; Neumann, 2010a; Sahrmann, 2002; Tortora & Nielsen, 2012). This anterior pelvic tilt accentuates lumbar lordosis (Neumann, 2010a), which can be potentially injurious and contribute to LBP as mentioned previously. In addition to the abdominal wall muscles, the ES muscles also play a major role in triplanar spinal stability, having major implications for lumbopelvic and postural control (McGill, 2007; Neumann, 2010a). The selected trunk muscles which were reviewed and examined within the present study are of particular interest when investigating the surface EMG of the lumbopelvic region (Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008; Nelson-Wong et al., 2013; Schinkel-Ivy & Drake, 2015), require further investigation within the insoles literature, and have implications for transient and chronic LBP (Dankaerts, O'Sullivan, Burnett, & Straker, 2006; Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2013).

2.2.2 Hip

The hip is a classic ball-and-socket joint which consists of the articulation between the head of the femur and the acetabulum (deep socket) of the pelvis (Drake et al., 2012; Neumann, 2010a). It serves as a central pivot point for the body as a whole, allowing simultaneous triplanar movements of the femur relative to the pelvis (flexion-extension, abduction-adduction, and internal-external rotation), as well as the trunk and pelvis relative to the femur (flexion-extension, lateral bend, and axial twist) (Neuman, 2010b). Proper functioning of the hips is especially important for the spine, as the pelvis acts as the platform for the spine, and a large amount of power is usually generated at the hips for both performance and safety reasons (McGill, 2007).With several major muscles (trunk, hip, and thigh) anatomically, and therefore functionally linked to it, the hip plays a dominant role in the alignment and movements across a large part of the body – especially the lumbar spine, pelvis, and femur (Neumann, 2010a; Tortora

& Nielsen, 2012). Taken together, the biomechanical functioning of the hip musculature has key implications for lumbopelvic and postural control.

While there are three major gluteal muscles (Drake et al., 2012), the gluteus medius (GTMED) is the primary focus of this review. The GTMED muscle is a fan shaped muscle, which has a broad origin on the superior, external part of the pelvis (the ilium), and attaches distally and laterally, on the greater trochanter of the femur (Drake et al., 2012; Tortora & Nielsen, 2012). It has three functional sets of fibers (anterior, middle, and posterior), with all of its fibers contributing to hip abduction, the anterior fibers producing hip internal rotation, and finally, the posterior fibers producing hip extension and external rotation (Clark & Haynor, 1987; Neumann, 2010a). The GTMED muscle is a powerful abductor of the hip, with the greatest abductor moment arm of all the abductor muscles, and the largest total abductor cross-sectional area (60%) (Clark & Haynor, 1987; Neumann, 2010a; Tortora & Nielsen, 2012). As a result of this function, the GTMED muscle is tremendously important for assisting the spine musculature (i.e. the obliques) in holding up the pelvis, and therefore reducing pelvic drop, during any activity requiring single leg stance (i.e. a OLST and gait) – this will be discussed subsequently in the context of the Trendelenberg test (see Section 2.3.2) (Drake et al., 2012; McGill, 2007). During single leg stance, the hip abductor muscles must produce a force twice that of body weight to stabilize the femoral head within the acetabulum, with the GTMED muscle in particular producing most of the compression force across the hip (Dalstra & Huiskes, 1995; Neumann, 2010a). Due to its connection between the lower extremity and lumbopelvic region, the GTMED muscle also plays a key role in stabilizing the pelvis and spine (McGill, 2007; Neumann, 2010a). This muscle was specifically targeted for the present study due to its major role in lumbopelvic and postural control within a LBP context (see Section 2.3.2) and due to its lack of investigation within the insoles literature.

2.2.3 Thigh

The thigh is the region of the lower limb consisting of the femur which proximally articulates with the pelvis (to form the hip joint) and distally articulates with the tibia (to form the knee joint) (Neumann, 2010a; Tortora & Nielsen, 2012). The function of the femur with respect to the pelvis has been described earlier when reviewing the hip joint. While there are several muscles of the thigh, only the biceps femoris (BF) and vastus medialis (VM) muscles will be reviewed. The BF muscle consists of two heads (focus here is on the long head), and it is the most lateral of the three hamstring muscles which make up the posterior compartment of the thigh (Drake et al., 2012; Tortora & Nielsen, 2012). Its long head crosses the posterior thigh obliquely from medial to lateral, spanning from the lower part of the pelvis (the ischial tuberosity), down to the lateral condyle of the tibia, and the head of the fibula (Drake et al., 2012; Tortora & Nielsen, 2012). In terms of its primary function, the long head of the BF muscle is a major hip extensor, and as a secondary function, it also adducts, and externally rotates the hip (Neumann, 2010a). Although the BF muscle is not a key lumbopelvic muscle, it contributes to pelvic and spinal stability due to its attachment to the pelvis which is linked with the lumbar spine (McGill, 2007; Neumann, 2010a). While the VM is not a lumbopelvic muscle, it is a proximal lower extremity muscle which may be of interest when examining the connection between the foot and lumbopelvic region. It consists of fibers that form two directions (proximally, the obliquus, and distally, the longus), and it is the most medial of the quadriceps muscles which make up the anterior compartment of the thigh (Neumann, 2010a; Tortora & Nielsen, 2012). Originating and inserting medially, it spans from the linea aspera of the femur, to the patella of the knee (via the quadriceps tendon), and the tibial tuberosity (via the patellar ligament) (Drake et al., 2012; Tortora & Nielsen, 2012). Its main function is knee extension, and it plays a major role in the stability of the patella during knee movements (Drake et al., 2012; Tortora & Nielsen, 2012). Due to its lack of connections with the pelvis and lumbar spine, the VM muscle does not contribute to pelvic or spinal stability. While the BF and VM muscles were the most distal muscles (relative to the lumbopelvic region) examined within the present study, and thus were not of primary interest, the BF was examined due its lower extremity and lumbopelvic connection, and the VM was examined due to its role in lower extremity movement and stability. With most of the insoles literature investigating the musculature of the shank, both of these more proximally located muscles (BF and VM) require further investigation within this literature.

2.2.4 Anatomy of the Trunk, Hip, and Thigh Summary

The trunk and hip muscles reviewed and examined within the present study play a crucial role in triplanar lumbopelvic and postural control, and can have major implications for LBP due their substantial contributions to spinal stability. Of these muscles, the GTMED was of primary interest due to its key role in the specific lumbopelvic and balance tests (particularly the OLST) examined within the present study. While the thigh muscles, particularly the VM, are not as critical in the context of this study, they play a major role in the movement and stability of the lower extremities, and may be of interest due to their position between the foot and lumbar spine. All of these muscles have been under-investigated within the insoles literature, and thus require further examination.

2.3 Functional Impairments, Clinical Tests, and Interventions of Lumbopelvic Control and Balance Associated with Low Back Pain

This section will first review major factors which contribute to motor control, followed by a targeted discussion surrounding impairments of lumbopelvic control, balance, and foot function within a LBP context. The theory and methodologies of the lumbopelvic and balance tests, and insoles used within the present study will be explained within the context of relevant literature.

2.3.1 Importance of Proprioception and Spinal Stability for Motor Control

While reduced physical fitness (Heneweer, Picavet, Staes, Kiers, & Vanhees, 2012) and physical inactivity (Heneweer, Vanhees, & Picavet, 2009) are associated with LBP, and thus often investigated, there has been an increased focus on the investigation of LBP within a motor control paradigm (Hodges & Moseley, 2003; O'Sullivan, 2005; Sahrmann, 2002). This requires an understanding of neuromuscular control, particularly, how the central nervous system (CNS) and musculoskeletal system interact to generate purposeful movements, through the coordination of individual muscles, joints, and limbs in relation to the rest of the body and the environment (Latash, Levin, Scholz, & Schöner, 2010; Shumway-Cook & Woolacott, 2007).

Optimal motor control involves the production of an ideal response based on the behavioural goal(s) and constraints associated with the specific motor task (Horak & Macpherson, 1996; Massion, 1992). Intact lumbar spine proprioception and sufficient spinal stability are of critical importance for this control, as the trunk plays a central role, supporting the limbs and their muscular attachments, and therefore affecting all other body segments through its position and function (Barwick, Smith, & Chuter, 2012; Henriksen, Lund, Bliddal, & Danneskiold- Samsøe, 2007; Sahrmann, 2002). As a result of its structural weakness (devoid of its supporting musculature), and its constantly competing demands (i.e. assisting with several physiological processes and its essential role in the challenges of postural control), it is not surprising that the motor control of the trunk can be compromised, leading to a multitude of issues, including LBP (McGill, 2007; Neumann, 2010a). While several factors can compromise trunk motor control, it is well documented that reduced lumbar spine proprioception (Brumagne, et al., 2000; Gill & Callaghan, 1998; Newcomer, Laskowski, Yu, Johnson, & An, 2000; O'Sullivan et al., 2003) and insufficient spinal stability (McGill et al., 2003; Reeves et al., 2011) are largely associated with these impairments in people with LBP.

Proprioception

Proprioception is the component of the somatosensory system which describes the sense of position and movement of one's own limbs and body without using vision (Gardner, Martin, & Jessel, 2000). Using different types of mechanoreceptors in muscles (muscle spindles to detect stretch, and Golgi tendon organs to detect contraction) and joints (to detect joint angles), proprioceptive mechanisms, especially in the trunk and foot (see Section 2.3.4), are crucial for whole body neuromuscular control (Gardner et al., 2000; Hidalgo, Gobert, Bragard, & Detrembleur, 2013). Thus, impairments in trunk proprioception can have deleterious effects on motor control as evident in individuals with LBP (Brumagne et al., 2000; Gill & Callaghan, 1998; Newcomer et al., 2000; O'Sullivan et al., 2003). This has been demonstrated in several studies which have observed higher lumbar spine repositioning errors (difference between the actual target position and the subject-perceived target position) in those with LBP compared to back-healthy controls during standing (Gill & Callaghan; Newcomer et al., 2000), four-point kneeling (Gill & Callaghan, 1998), and sitting tasks (Brumagne et al., 2000; O'Sullivan et al., 2003). Beyond the proprioceptive deficits in the lumbar muscles and joints, these researchers suggested that pain may have also contributed to these motor control impairments. Although it is unclear whether these proprioceptive deficits are causal or compensatory with regard to LBP (Brumagne et al., 2000), they can have major implications in terms of lumbopelvic, whole body postural control, and LBP.

Spinal Stability

Whole body stability (balance) refers to the dynamics of body posture to prevent falling (Winter, 1995a). However, spinal stability refers to the mechanical stability of the spine which is achieved through the passive (ligamentous), active (musculotendenous), and neural control subsystems (Panjabi, 1992). As EMG activity was one of the main outcome variables in the present study, the role of the active musculotendenous subsystem will be emphasized throughout this review. Without its associated musculature, the osteoligamentous spine alone is extremely unstable, buckling under just ~20 lbs of compressive load (Lucas & Bresler, 1961). Thus, muscles play a crucial role in creating a balanced stiffness and resulting stability of the spine, in all of its degrees of freedom (three rotational and three translational nodes), at each intervertebral joint (McGill et al., 2003). The achievement of optimal spinal stability is quite complex and an important component of lumbopelvic control and balance (McGill et al., 2007; Reeves et al., 2007). It involves highly coordinated muscle activation patterns from the full complement of stabilizing musculature (i.e. abdominal, back, and hip muscles), which must continually adapt to the specific demands of the task (McGill et al., 2003). The inappropriate activation of just one muscle (magnitude and/or timing) can produce instability (McGill et al., 2007).

Co-contraction refers to the simultaneous activation of opposing or antagonistic muscles (Chiou, Lee, & Chen, 1999). It is an important aspect of spinal stability and particularly, maintaining neutral upright postures, even in a healthy spine (Granata & Marras, 2000; Silfies, Squillante, Maurer, Westcott, & Karduna, 2005; Van Dieen, Cholewicki, & Radebold, 2003).

However, this co-activation of stabilizing muscles involves an important trade-off in terms of stability, metabolic cost, and injury potential (McGill et al., 2003; Reeves et al., 2011). For instance, very moderate levels of abdominal wall activation (~10% of MVC: maximum voluntary contraction) are required to perform most tasks of daily living (McGill, 2007). Co-contraction which is in excess of what is required for a particular movement may induce unnecessary increases in spinal loads which has shown to be associated with low LBP (Granata & Marras, 2000; Hodges & Richardson, 1996). In these situations, an attempt at increasing spinal stability occurs at the cost of increased spinal loading (Granata & Marras, 2000). There is evidence of increased trunk muscle co-contraction, and impaired postural control in individuals with LBP (Jones, Henry, Raasch, Hitt, & Bunn, 2012). This can be understood in terms of a trunk stiffening strategy which has been observed in individuals with LBP as a protective mechanism to try and prevent pain (see Section 2.3.3). Specifically, it has been observed as a potential causal factor in transient LBP development during prolonged standing (Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2008), and as a potential compensatory response to clinical LBP during sitting (Dankaerts et al., 2006) and dynamic lifting tasks (Marras, Davis, Ferguson, Lucas, & Gupta, 2001). Thus, while a sufficient level of spinal stability is an important component of postural control, when excessive, it can be detrimental for this control, and potentially injurious as evident in LBP populations.

2.3.2 Lumbopelvic Control Impairments, Tests, and Low Back Pain

Lumbopelvic Impairments

Optimal lumbopelvic control involves the ideal functioning and interaction between the musculature of the lumbar spine, pelvis, and hip as exhibited through controlled and aligned postures and movements, based on the specific task and environment (Henriksen et al., 2007;

Sahrmann, 2002). In addition to intact lumbar spine proprioception (Gill & Callaghan, 1998; Newcomer et al., 2000; O'Sullivan et al., 2003), to achieve this optimal lumbopelvic control, the muscles of this region must sufficiently activate (magnitude and timing) to generate the ideal amount of stiffness for spinal stability, while still facilitating the motion required for the postural control task (McGill et al., 2003; Neumann, 2010a). Lumbopelvic control is a central component of whole body postural control during many postures and movements including upright standing and gait (Sahrmann, 2002; Thorstensson et al., 1982). Thus, any dysfunction in this control can largely impair functional activities of daily living and lead to LBP development.

While most of the LBP literature focuses on the musculature of the trunk (Dankaerts et al., 2006; Marras et al., 2001; Newcomer et al., 2002; Silfies et al., 2005), the hip musculature, particularly the GTMED muscle, also appears to play an important role in lumbopelvic control and LBP (Marshall et al., 2011; Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2008). The GTMED muscle's role in lumbopelvic control and LBP is not surprising due to its functional connections with the pelvis and spine via the lower extremity, and its resulting impact on movements across a large part of the body (Neumann, 2010a, Tortora & Nielsen, 2012).

The neuromuscular control of the GTMED muscle consistently appears to have an important role in transient LBP development during prolonged standing (Marshall et al., 2011; Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008). Using a two hour prolonged standing occupational simulation, all of these researchers observed significantly higher bilateral GTMED co-contraction in PDs compared to NPDs, with NPDs demonstrating reciprocal firing of the right and left (R/L) GTMED muscles. Based on the single factor of GTMED muscle co-contraction alone, Nelson-Wong and colleagues (2008), and Marshall and colleagues (2011) were able to correctly predict the presence (PD group) or absence (NPD group) of a clinically relevant level of transient LBP in 74% and 79% of participants, respectively. What was of particular note was the presence of this co-contraction prior to the onset of perceived LBP development. This indicated that this GTMED muscle activation pattern may be a causal factor for this LBP. Coupled with the presence of trunk flexor-extensor co-contraction observed in PDs during prolonged standing, Nelson-Wong & Callaghan (2010a) postulated that this bilateral GTMED co-activation may reflect an inability to control trunk and abdominal muscles for effective lumbopelvic motor control during prolonged standing. Taken together, these studies emphasize the importance of the GTMED as a predictive muscle in the development of transient LBP development within a motor control paradigm.

Lumbopelvic Control Tests

Movement control tests by definition, assess the ability of muscles to isometrically hold a position or prevent motion at one joint, while concurrently producing an active movement at another joint (Mottram & Comerford, 2008). The ability to actively control and prevent compensatory movement when required or instructed to do so is considered controlled motion, while the inability to do this is considered uncontrolled motion (Mottram & Comerford, 2008). Lumbopelvic movement control tests, in particular, examine the ability of an individual to control and reposition the lumbopelvic complex when challenged in different directions (Grosdent et al., 2015). Given the criteria for controlled and uncontrolled motion above, when considering the prone lying active knee flexion test for example, correct performance involves at least 90° of active knee flexion without rotational movement of the low back and pelvis; conversely, incorrect performance involves pelvis rotation with active knee flexion (Luomajoki, Kool, de Bruin, & Airaksinen, 2007).

While the literature is mixed, a number of studies demonstrate evidence of associations between worse performance on lumbopelvic control tests and LBP and/or injury (Babiolakis et al., 2015; Luomajoki et al., 2007; Nelson-Wong et al., 2009; Roussel et al., 2009). When examining ten lumbopelvic control tests (one sitting, four standing, two prone, one supine, and two four-point kneeling), Luomajoki and colleagues (2007) observed worse performance (2/3 more tests performed incorrectly) in LBP patients compared to back-healthy controls. However, these researchers did not identify which tests LBP patients tended to perform worse in. Roussel and colleagues (2009) assessed the ability of four commonly used lumbopelvic control tests (one standing and three supine) to predict lower limb and/or lumbar spine injuries in dancers. Only two of the four tests (one standing and one supine) were found to be predictive of lower extremity or lumbar spine injuries in the dancers, correctly allocating 78% of the dancers into the injury group.

The Active Straight Leg Raise (ASLR) test, a supine test of lumbopelvic control test was one of the lumbopelvic control tests examined by Roussel and colleagues (2009) which was not successful in predicting lower limb and/or lumbar spine injury. In line with Roussel et al., 2009, Nelson-Wong et al. (2009) and Babiolakis et al. (2015), also found the ASLR test to be unsuccessful in its association with LBP and back injury. However, these studies included a novel lumbopelvic movement control test, the AHAbd test, which was a major clinical test of interest in the present study. The AHAbd test is a reliable observational screening tool (Davis, Bridge, Miller, & Nelson-Wong, 2011) which emphasizes frontal plane trunk control and is heavily based on the identification of the GTMED muscle as predictive of LBP development during standing (Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008). It assesses lumbopelvic control during a hip (GTMED muscle) initiated active lower limb movement which is performed in an inherently unstable sidelying position (Nelson-Wong et al., 2009). Out of 17 standardized clinical measures typically assessed with LBP patients, the AHAbd test was the *only* clinical assessment measure to distinguish between PDs and NPDs, with significantly worse lumbopelvic control (higher scores on an ordinal scale of 0-3: no – severe loss of pelvis frontal plane position) observed in PDs (mean ± standard deviation (SD): 1.35 ± 0.93 vs. 0.65 ± 0.75). These measures ranged from passive trunk and hip range of motion tests, to tests of side-support endurance, and lumbar segmental mobility (Nelson-Wong et al., 2009). To add to the strength of this finding, individuals receiving a failed AHAbd test score (≥2 moderate loss of pelvis frontal plane position, reflective of reduced lumbopelvic control) were based on an odds ratio, 3.85 times more likely to develop transient LBP during prolonged standing (Nelson-Wong et al., 2009).

Babiolakis and colleagues (2015) observed a similar result also using a comprehensive protocol. The AHAbd test was the *only* performance-based test out of 22 physical fitness and biomechanical measures to distinguish between nurses with and without a recent history of LBP due to back injury (Babiolakis et al., 2015). There was a wide variety of measures in this study with biomechanical tests ranging from a modified deep squat, neck, shoulder, trunk and hip active range of motion, to ground reaction force measures during quiet standing, and physical fitness tests of muscular strength, endurance and flexibility (Babiolakis et al., 2015). While the AHAbd test was the only test to distinguish between groups, Babiolakis et al. (2015), only observed a difference when examining the left leg AHAbd test outcomes, with recently injured nurses demonstrating reduced lumbopelvic control (higher AHAbd test scores) compared to not recently injured nurses (1.6 ± 0.5 vs. 1.2 ± 0.4).

The examination of this test in both of these studies was observational, based on rated scores. However, in 2013, Nelson-Wong and colleagues assessed this test using surface EMG to compare the muscular activation patterns of individuals with LBP and back-healthy controls (no LBP that was significant enough to seek medical care of that resulted in greater than three days off from work, school, or recreation within the past five years). In line with the observational work, individuals with LBP exhibited worse lumbopelvic control (more variable muscle activation patterns) compared to back-healthy individuals who exhibited better lumbopelvic control (a more consistent proximal to distal activation). Thus, lumbopelvic control assessed by the AHAbd test has been found to reveal both visually identified movement control differences and EMG quantified motor control differences between individuals with and without transient/clinical LBP. The present study was also interested in examining both observational and EMG quantified lumbopelvic control during this test. Taken together, these studies highlight the unique and important ability of the AHAbd test to detect lumbopelvic impairments in individuals with transient or a recent history of LBP in protocols where numerous other performance-based measures did not (Babiolakis et al., 2015; Nelson-Wong et al., 2009).

Unlike all other tests of lumbopelvic control which are performed in more stable (i.e. prone, supine, four-point kneeling, and standing positions (Babiolakis et al., 2015; Luomajoki et al., 2007; Nelson-Wong et al., 2009; Olivier et al., 2015; Roussel et al., 2009), the AHAbd test is unique in that it is performed in an inherently stable sidelying position. Given its moderately strong links to LBP development during standing (Nelson-Wong et al., 2009), and as the only measure out of several typically used clinical measures to demonstrate this link in previously asymptomatic (back-healthy) individuals, this test appears to be particularly sensitive to detecting lumbopelvic impairments in what can be considered a ‘sub-clinical’ (currently asymptomatic for LBP but at risk of future LBP) population (Nelson-Wong et al., 2010a). However, a notable limitation of the AHAbd test is its use of a non-weight-bearing posture which is not functionally relevant. Thus, if an association between this sidelying test and upright balance tests (i.e. OLST, SRT) which are also heavily based on lumbopelvic control was observed, these more functionally relevant tests could be used as additional, related clinical screening tools for LBP development. While several lumbopelvic control and a few standing balance tests have both been examined in the same study (Olivier et al., 2015), no previous studies, to the author's knowledge, have examined the association between these tests. What is interesting to note in Olivier and colleagues' (2015) study is that none of the nine lumbopelvic control tests (one of which included an eyes open, firm surface OLST) were predictive of low back (and lower extremity) injury in this asymptomatic (back-healthy) elite cricket athlete population, yet, two posturally challenging single-limb stance balance tests (one of which was included an eyes closed, compliant surface OLST) were predictive. Given the relationship between lumbopelvic control and standing balance tests (i.e. the OLST) with LBP and/or injury separately, the examination of the association between the AHAbd test and standing balance tests was of interest in the present study.

Similar to the AHAbd test, the Trendelenberg test is another lumbopelvic control test which also focuses on the hip abductors in the context of LBP. However, this test is performed in an upright standing, more functionally relevant position. This test is a common clinical assessment of the lower back, pelvis, and hip which is used to evaluate an individual's posture during single leg stance (Kendall et al., 2013; Whatling, Holt, & Beynon, 2015). This test specifically evaluates the ability of the hip abductors (i.e. GTMED muscle) to stabilize the pelvis on the femur during the transfer of load when assuming a single leg stance position (Kendall et al., 2013; Kendall et al., 2010; Magee, 2007). Normally, the pelvis on the contralateral (non-stance leg) side should rise, keeping the pelvis horizontal; this reflects a negative test (Kendall et al., 2013; Magee, 2007). However, a positive test is identified when the pelvis on the contralateral side drops below the horizontal position during the transition to single leg stance (Magee, 2007). This test is typically administered by clinicians, who assess the presence or absence of pelvic drop based on palpation and observation, which does not include a quantification of the magnitude of the movement. Pelvic drop on the contralateral side observed during a positive test is suggested to be caused by weakness of the GTMED muscle on the ipsilateral (stance leg) side (Magee, 2007). This pelvic drop may also be accompanied by other compensations which have been postulated to reflect greater weakness of the hip abductors (Sahrmann, 2002). These can include trunk flexion towards the ipsilateral side, and hip adduction and internal rotation of the contralateral side (Kendall et al., 2010; Sahrmann, 2002).

When reviewing the literature, there is a lack of research which examines pelvic drop. While clinicians commonly use this test based on the postulation that hip abductor (i.e. GTMED muscle) weakness is associated with pelvic drop, and likely contributes to hip or LBP (Kendall et al., 2010), the empirical evidence is lacking and does not support this association (Penney, Ploughman, Austin, Behm, & Byrne, 2014). While no association between GTMED muscle weakness and pelvic drop was observed in these studies (DiMattia, Livengood, Uhl, Mattacola, & Malone, 2005; Fetto, Leali, & Moroz, 2002; Kendall et al., 2013; Kendall et al., 2010), neuromuscular control differences in the GTMED muscle were observed by Penney and colleagues (2014) who observed a greater activation of the GTMED in chronic LBP patients compared to back-healthy controls during the transition from double to single-leg stance. Thus, further investigation of pelvic drop, which includes neuromuscular control measures of the GTMED muscle, is required to clarify its association with LBP.

2.3.3 Balance Impairments, Tests, and Low Back Pain

Balance Impairments

Unlike spinal stability which involves the coordination of the trunk muscles to produce an adequate amount of stiffness (McGill, 2007), postural control involves the complex and dynamic interaction between sensorimotor processes to achieve two main functional goals, postural orientation and balance (Horak, 2006; Shumway-Cook & Woolacott, 2007). Postural orientation refers to the relative positioning of the body segments with respect to each other and the environment, while balance, describes the dynamics of body posture to prevent falling, and is a critical component of most tasks (Horak & Macpherson, 1996; Winter, 1995a). To achieve postural control, inputs from three main sensory systems, somatosensory (proprioceptive and cutaneuous information relative to the support surface), vestibular (gravity), and visual (relationship of body to objects in the environment) are continuously integrated by the CNS, and the musculoskeletal system then uses this information to generate appropriate muscular responses (Horak & Macpherson, 1996; Shumway-Cook & Woolacott, 2007). Based on the immediate conditions (the specific task and environment), the CNS must dynamically weight these sensory inputs relative to one another to maintain balance (Brumagne, Cordo, & Verschueren, 2004; Carver, Kiemel, & Jeka, 2006; Horak & Macpherson, 1996). By selectively challenging these sensory systems (i.e. using an eyes closed condition and tests with a small/narrow BOS), researchers can study their contribution to balance.

Without a postural control system continuously acting, humans are considered an inherently unstable system (Winter, 1995a). During upright standing, humans have a small BOS, a high centre of mass (COM), and two-thirds of their body mass (head, arms, and trunk) is located two-thirds of body height above the ground (Horak & Macpherson, 1996; Winter, 1995a). Thus, to remain balanced, by keeping or returning the COM over its BOS, the force of gravity must constantly be counteracted with muscular responses (Horak, 1987). To understand this dynamic process of balance in terms of the biomechanical variables which are commonly used to quantify it, a brief description of the COM, COP, and their relationship is required. The COM of the body refers to the point location at which the entire mass of the body is balanced, and it represents the weighted average of the COM of each body segment in three-dimensional space (Horak & Macpherson, 1996; Winter, 1995a). With changes in postural orientation, the COM can experience large changes in position as it is considered a passive variable which is guided by the postural control system (Horak & Macpherson, 1996; Winter, 1995b). Conversely, the COP is the point location of the vertical ground reaction force vector (Winter, 1995a). The COP acts as the neuromuscular regulator of the COM position within the BOS, as it moves with the generation of active muscular forces to control imbalances in the position of the COM (Palmieri, Ingersoll, Stone, & Krause, 2002; Winter, 2009). To maintain balance during shifts in COM position, the COP excursions are always greater than the COM, with the COP oscillating on either side of the COM at a higher frequency and greater amplitude (Palmieri et al., 2002; Winter, 1995a). For instance, when a posterior displacement of the COM is detected by the CNS, a reflexive contraction of the dorsiflexors will occur, moving the COP posterior to the COM in an effort to regain balance (Palmieri et al., 2002). While it is evident that the COM and COP are closely related, their trajectories are completely independent (Winter, 2009). It is critical to be aware of this distinction as many researchers have mistakenly treated the COM and COP as synonymous terms, inaccurately referring to the COP as body ‘sway’ (Winter, 1995a, 2009).

It is also important to understand the two main sagittal plane (A-P) postural control strategies which are commonly used to maintain balance during bilateral stance (Horak & Nashner, 1986; Winter, 1995a). The ‘ankle’ and ‘hip’ strategies represent extremes on a continuum, with normal postural control tending to include elements of both in varying proportions depending on the postural context (Horak & Nashner 1986; Mok et al., 2004). The ‘ankle strategy’, which is the most commonly used response during quiet standing, involves a distal to proximal muscle activation sequence in which the dominant torque is generated by the plantar- and dorsi-flexors at the ankle, and minimal movement occurs about the knee and hip joints to maintain the COM over the BOS (Horak, 1987, 2006; Horak & Nashner, 1986; Winter, 1995a). This strategy is appropriate to maintain balance for slow and small perturbations when standing on a firm, wide, and even surface (Horak, 1987, 2006; Horak & Nashner, 1986). Conversely, a ‘hip strategy’ involves flexion or extension at the hips where a proximal to distal muscle activation sequence is employed, in which the horizontal shear forces against the surface are largely generated from torques at the hip (Horak, 1987, 2006; Horak & Nashner, 1986). This strategy is typically used for large and fast perturbations (where ankle torques are insufficient to shift the COM) when standing in a narrow (i.e. tandem, heel-to-toe) stance, on a compliant, narrow, and uneven surface (Horak, 1987, 2006; Horak & Nashner, 1986).

As discussed previously (see Section 2.3.1), intact proprioception is a critical component of optimal movement control (Silfies, Cholewicki, Reeves, & Greene, 2007). Thus, deficits in proprioception have been strongly suggested to cause balance impairments in individuals with LBP (Brumagne, Janssens, Janssens, & Goddyn, 2008; Mok, Brauer, & Hodges, 2004) as evidenced by a reduced sense of their lumbar spine position in space (Brumagne et al., 2000; Newcomer et al., 2000; O’Sullivan et al., 2003; Sheeran , Sparkes, Caterson, Busse-Morris, &

van Deursen, 2012). These proprioceptive deficits have been postulated to alter the neuromuscular control of these individuals, disturbing their balance between movement and stability (Massion, 1994). This is evidenced when individuals with LBP use a dysfunctional (excessive) trunk stiffening strategy during upright standing during conditions of postural anxiety (Brumagne et al., 2008; Jones et al., 2012; Mok et al., 2004). Specifically, as a preparatory strategy for an upcoming perturbation (postural threat), these individuals exhibit an increased baseline level of trunk muscle co-contraction likely intended to restrict trunk movement due to their anticipation or fear of impending pain which they believe may result from large movements or accelerations of the trunk (Jones et al., 2012). With a reluctance to use a hip strategy which would lead to these feared movements and accelerations of the trunk, individuals with LBP inappropriately rely more heavily on an ankle strategy when a hip strategy should dominate during fast and large perturbations, and/or on a compliant, narrow, and uneven surface (Braga et al., 2012; Henry, Hitt, Jones, & Bunn, 2006; Jones et al., 2012). This strategy appears to be counterproductive in that it exposes the spine to greater compressive forces and a greater resultant displacement which is what these individuals were trying to prevent (Brumagne et al., 2008; Cholewicki, Simons, & Radebold, 1997; Jones et al., 2012; Mok et al., 2004, 2007). Overall, this dysfunctional postural control strategy actually leads to more frequent losses of balance in individuals with LBP, and may contribute to injurious, rather than protective neuromuscular control (Brumagne et al., 2008; Mok et al., 2004).

Balance Tests

Human balance can be examined using several different tests and under many different conditions. Researchers can manipulate the postural context to target specific sensory information, forcing the CNS to reweight available sensory demands based on the current task and environment (Brumagne et al., 2004; Carver et al., 2006; Horak & Macpherson, 1996). For instance, normal conditions may involve a quiet standing position, with eyes open, standing on a firm surface. However, to increase the demands on the somatosensory system, researchers can remove visual input (using an eyes closed condition), and can use a test which requires a small/narrow BOS (Horak & Macpherson, 1996). One of the simplest assessments of balance includes recording the length of time participants can maintain a particular equilibrium position in normal and altered conditions (Horak, 1987). While maximum balance times are very informative for certain populations (i.e. fall risk in older adults; Hurvitz, Richardson, Wernet, Ruhl, & Dixon, 2000), they can be somewhat less informative during an eyes open condition in a young, non-clinical population that may already be performing near optimal balance levels (Springer, Marin, Cyhan, Roberts, & Gill, 2007). However, when visual input is removed, young non-clinical individuals tend to achieve markedly lower standing balance times (Springer et al., 2007) which when compared to an eyes open condition, can provide information regarding the roles of visual and somateosensory inputs as will be discussed further in the context of Romberg quotients and relative change equations.

Beyond assessing balance using just timing information, more sophisticated biomechanical measures can be used. The COP is very commonly used as a measure of standing balance using one or two force platforms. In a young, non-clinical population, a small amplitude COP displacement (COP_d) tends to reflect ‘good’ control of balance, while a higher displacement amplitude reflects ‘poor’ control (Braga et al., 2012). An increased magnitude of the COP_d is assumed to reflect a reduced ability to control the COM, as a greater COP_d implies an increased neuromuscular response to compensate for an increased displacement in the COM; the converse is assumed to be true for a decreased or small COP_d (Palmieri et al., 2002). However, depending on the population that is being studied, this is not always the case as evident in parkinsonian (Horak, Nutt, & Nashner, 1992) and spinal-cord injured (Karataş, Tosun, & Kanatl, 2008) populations where a small COP_d was indicative of reduced postural control. Balance may also be examined using EMG and kinematic measurements to identify whether individuals exhibit healthy, altered, or impaired (i.e. trunk stiffening in people with LBP) postural control strategies.

Two widely used and easily administered clinical balance tests of double-stance include the classical Romberg test, and its modification, the SRT (which was examined within the present study). The classical Romberg test, developed around 1851-1853 (Dornan, Fernie, & Holliday, 1978; van Griethuysen, Paul, Andrews, & Nicol, 1982), is a static upright standing balance test which requires individuals to stand with their feet close together (Briggs, Gossman, Birch, Drews, & Shaddeau, 1989; Nelson-Wong et al., 2012). This test was originally designed to examine the amount of body sway exhibited by individuals with posterior column disorders (Briggs et al., 1989; van Griethuysen et al., 1982). The SRT, is more challenging and sensitive than the classical Romberg test as it involves a reduction in somatosensory input using a narrow (tandem heel-to-toe) BOS (Findlay, Balain, Trivedi, & Jaffray, 2009; Horak, 1987; Nelson-Wong et al., 2012; Rogers, Takeshima, & Islam, 2003; Steffen & Seney, 2008). The use of a narrow tandem stance BOS also means that this test is heavily reliant on lumbopelvic control through a hip dominant strategy, as opposed to an ankle dominant strategy (Horak, 1987, 2006; Horak & Nashner, 1986). While the BOS for the SRT is larger in the sagittal plane (A-P axis) than the OLST, it is considered to be a valuable alternative to the single-leg stance test when one is trying to increase the demands of the postural control system in the frontal plane specifically (Palmieri et al., 2002).

The SRT can be performed with several variations including: eyes open/closed (Heitmann, Gossman, Shaddeau, & Jackson, 1989); arms free to move (Briggs et al., 1989; Heitmann et al., 1989; Steffen and Seney 2008) or folded across the chest (Johnson et al., 2005; Newton, 1989); and with the dominant foot positioned behind the non-dominant foot (Briggs et al., 1989; Heitmann et al., 1989; Steffen & Seney, 2008), or not specified (Johnson et al., 2005; Newton 1989). Beyond reaching the maximum test time (typically 30/60 seconds), and movement of the arms from a specified posture (when this applies), the termination criteria for the SRT differs between studies when considering the foot placement: this ranges movement of the feet from the proper position (Briggs et al., 1989; Heitmann et al., 1989; Johnson et al., 2005; Steffen & Seney, 2008), to a loss of balance exhibited through taking a step or falling (Nelson-Wong et al., 2012; Newton, 1989; Rogers et al., 2003). The SRT has been assessed within a number of different populations including older adults (Briggs et al., 1989; Gschwind et al., 2013; Heitmann et al., 1989) divers recovering from decompression sickness (Fitzgerald, 1996; Lee, 1998), individuals with mild acute mountain sickness (Johnson et al., 2005), and in patient populations (Newton 1989; Steffen & Seney, 2008). However, to the author’s knowledge, it has not been assessed within a LBP population.

Similar to the SRT, the OLST is also a widely used, easy to administer clinical balance test which assesses static balance maintenance in an upright standing posture (Maribo et al., 2009; Rogers et al., 2003). The OLST also involves reduced somatosensory input (Horak, 1987), however, it is considered more difficult and more sensitive than the SRT as it involves a smaller BOS in both the sagittal (A-P axis) and frontal (M-L axis) planes (Palmieri et al., 2002; Rogers et al., 2003). It is also a balance test which heavily relies on lumbopelvic control, due to its narrow and small BOS (Horak, 1987, 2006; Horak & Nashner, 1986; Palmieri et al., 2002), and during the transition from double to single-leg stance, when the GTMED muscle plays a crucial role in maintaining the level of the pelvis as mentioned previously (Drake et al., 2012; McGill, 2007).

There is no consensus within the literature as to how the OLST is performed (Maribo et al., 2009). There are multiple variations, with the test being performed with: eyes open (Jonsson, Seiger, & Hirschfeld, 2004; Maribo et al., 2009; Vereeck, Wuyts, Truijen, & Van De Heyning, 2008) and closed (Maribo et al., 2009; Vereeck et al., 2008); with (Ageberg, Roberts, Holmström, & Fridén, 2003; Maribo et al., 2009) and without (Vereeck et al., 2008) visual fixation; with free (Jonsson et al., 2004; Vereeck et al., 2008) and specified (Maribo et al., 2009) arm postures; barefoot (Ageberg et al., 2003; Maribo et al., 2009) and shod (Nelson-Wong et al., 2012); with standardized (Ageberg et al., 2003; Maribo et al., 2009) and non-standardized (Nelson-Wong et al., 2012) positions of the feet during the single and double-support phases; and for varying durations of time typically ranging from 30-60 seconds (Jonsson et al., 2004; Maribo et al., 2009; Nelson-Wong et al., 2012; Vereeck et al., 2008). The initial position of the OLST involves a relaxed bilateral stance with weight evenly distributed between both feet, which is followed by a dynamic transition phase into single-stance where individuals attempt to maintain balanced in this static upright standing posture until any of the termination criteria are met (Jonsson et al., 2004). While the literature often is not explicit in terms of when the test officially begins, in the studies which do specify a definitive starting point, this appears to occur at the beginning of the single-stance phase (Briggs et al., 1989; Jonsson et al., 2004; Rogers et al., 2003; Vereeck et al., 2008). Similar to the SRT, the end of the test is marked by slight differences in the termination criteria beyond reaching the maximum test time. Some of these differences are a result of the varying postural requirements of the test; however, some of the differences involve the components of the test which are common amongst the different versions. Some studies were specific with their termination criteria, ending when the participant disengaged from the starting position: when the swing leg (leg being lifted) touched the support surface (Briggs et al., 1989; Jonsson et al., 2004; Rogers et al., 2003); when the stance (weight-bearing) leg was displaced (Briggs et al., 1989; Rogers et al., 2003), or when the foot of the swing leg was used to support the stance leg (Briggs et al., 1989; Rogers et al., 2003). Other studies were more ambiguous with their criteria, with the test ending when the participant lost their balance (Maribo et al., 2009; Nelson-Wong et al., 2012).

The OLST has been used in a variety of research and clinical settings, however, only a few studies have investigated differences in balance between individuals with chronic LBP and back-healthy controls during balance tests of single-support (da Silva et al., 2015; Ham et al., 2010; Luoto et al., 1998). Despite the different methodologies employed and outcome measures examined (i.e. kinematic, and kinetic COP), all of these studies observed significantly worse balance in individuals with chronic LBP compared to the back-healthy controls. These researchers postulated that these balance impairments in affected individuals may be due to several reasons including reduced trunk proprioception and lumbopelvic motor control which may be more evident and occur more easily in single-limb vs. double-limb standing tasks (da Silva et al., 2015; Ham et al., 2010; Luoto et al., 1998).

When assessing both eyes closed and eyes open balance tests, the eye conditions can be analyzed separately or in combination, as a relative comparison. This comparison has been performed in previous work using a Romberg quotient (eyes closed/eyes open) to compare balance measures such as the COP (Bronstein, Hood, Gresty, & Panagi, 1990; Lê & Kapoula, 2008; Menegoni et al., 2011; Morioka, Okita, Takata, Miyamoto, & Itaba, 2000). In the present study, it was also of interest to quantify balance outcomes (EMG, kinematic, and kinetic, and balance time) with a comparison of eyes closed and eyes open conditions using the following relative change Equation 1 (Dill & Costill, 1974; Perry et al., 2015):

Equation 1: $$\left(\frac{\text{Eyes closed} - \text{Eyes open}}{|\text{Eyes open}|}\right) * 100$$

The interpretation of the magnitude resulting from this equation is the same for the Romberg quotient and relative change equation. A higher number reflects a greater dependence on visual information and by extension, a decreased balance performance in the absence of vision, while the converse is true for a lower number (Bronstein et al., 1990). A value of zero represents no difference between the two visual conditions (Losa Iglesias et al., 2012). The importance of its use within the present study results from its ability to allow researchers to indirectly discriminate between different sensory inputs, with a particular emphasis on the proprioceptive system (Menegoni et al., 2011). For non-clinical populations, the eyes open condition involves the integration of visual, vestibular, and somatosensory (cutaneous and proprioceptive) information, while during the eyes closed condition, only vestibular and somatosensory information can be relied upon (Menegoni et al., 2011). Thus, in the absence of vision, somatosensory inputs become critical in the context of postural control (Horak & Macpherson, 1996). In relation to the insoles used in this study, enhancements in proprioception which may result from the use of these devices were expected to improve balance performance. Improvements were particularly expected during the eyes closed condition which is dominated by somatosensory inputs as would be reflected in a relative change magnitude that is closer to zero.

2.3.4 Foot Impairments, Insoles, and Low Back Pain

Foot Impairments

The human foot is the first contact between the body and the external environment and therefore serves a critical role in upright standing and locomotion (Chen, Chou, Tasi, Lo, & Kao, 2014; Wilson, Rome, Hodgson, & Ball, 2008). The functioning of the foot is both important and complex, as it is postulated to have major implications for lumbopelvic control, whole body postural control, and LBP (Barwick et al., 2012; Bird & Payne, 1999; Christovão et al., 2013). Within a LBP context, the proximal effects of the foot will be explained through mechanisms of abnormal foot pronation, inadequate shock absorption, and proprioception.

Abnormal foot structures and functions have been widely recognized as a cause of altered alignment, mechanics, and concomitant pain and injury of the foot, ankle, knee, hip, and back (Barwick et al., 2012; Kendall, Bird, & Azari, 2014; Riskowski et al., 2013). In terms of foot function, while both abnormal pronation (which flattens the medial longitudinal arch of the foot) and supination (which raises the medial longitudinal arch of the foot) (Kirby, 2000) have been associated with cumulative overuse injuries, most of the literature tends to focus on pronation (McClay & Manal, 1998; Nigg, Cole, Nachbauer, 1993). Beyond extreme foot structures, weakness and abnormal mechanics of the structures which support and control the arch have been postulated to be causal factors of this dysfunctional pronation (Neumann, 2010a).

While the underlying mechanisms are complex and not fully understood, one of the ways to understand how dysfunction at the foot can cause mechanical LBP, is through the abnormal functioning of the foot during the stance phase of gait. The normal functioning of the foot must first be described to serve as a basis for comparison. At the beginning of the stance phase, which is identified by a heel strike, the subtalar joint should be functioning from its neutral position; this has been described as one-third of the total range of motion of the subtalar joint from the full everted position (Donatelli, Hulbert, Conaway, & Pierre, 1988; Root, Orien, Weed, & Hughes, 1977). Immediately after heel strike, the subtalar joint pronates, flattening the arch of the foot, to act as a shock absorbing mechanism and to allow the foot to adapt to ground surfaces (Donatelli et al., 1988; Kirby, 2000). Towards the end of stance phase (from midstance to toe-off), the subtalar joint supinates, to form a rigid lever for propulsion during the push-off phase of gait (Donatelli et al., 1988; Kirby, 2000). During abnormal functioning, there is an excessive, prolonged, or poorly controlled pronation of the subtalar joint during the stance phase which prevents the timely (earlier) occurrence of supination required to stabilize the midfoot for push-off (Castro-Méndez, Munuera, & Albornoz-Cabello, 2013; Donatelli et al., 1988; Neumann, 2010a).

As a result of this abnormal foot pronation, it has been suggested that these mechanics of the foot are transmitted proximally up the kinematic chain to the lumbopelvic region (Barwick et al., 2012; Bird & Payne, 1999). With respect to LBP, this mechanism is proposed to occur through excessive internal rotation of the tibia and femur, anterior pelvic tilt, and lumbar lordosis which alter the normal biomechanics of the lower back, increasing the strain on lumbopelvic joints and muscles (Barwick et al., 2012; Bird & Payne, 1999; Botte, 1981; Kendall et al., 2014; Rothbart, Hansen, Liley, Yerrart, 1995; Tiberio, 1987, 1988). Increased strain on pelvic muscles including the iliopsoas, piriformis, and gluteus maximus may lead to rotational stress on the lumbar vertebra and associated sacroiliac and lumbosacral instability (Bird & Payne, 1999; Michaud, 1997). Rotational stress in particular is not well tolerated by the lumbar spine and is associated with disc weakening and injury (Danaberg & Giuliano, 1999; Farfan, Cossette, Robertson, Wells, & Kraus, 1970; Marras & Granata, 1995).

The link between abnormal foot pronation and LBP can also be understood in terms of impact-related pathology. During heel strike, an impulsive shock wave travels from the lower extremities, through the spine, up to the head (Forner et al., 1995). In just 20-80 milliseconds, the pressure resistance force acting on the foot increases to 1.5 times body mass (Wosk & Voloshin, 1981). Thus, even under normal conditions, the accumulation of shock waves generated with human motion is proposed to cause a progressive weakening of shock absorbing structures and mechanisms (Voloshin & Wosk, 1982). Using both active and passive mechanisms (Oakley & Pratt, 1988), appropriate subtalar joint pronation and knee flexion at heel strike are some of the ways by which the body works to attenuate the potentially damaging effects of this shock wave (Donatelli et al., 1988; Lockard, 1988). When these mechanisms are intact, absorption is suggested to be successfully reduced from 50-90% of the heel strike by the time the shock wave reaches the knee, and by up to 98% when it reaches the head (Forner et al., 1995). However, with dysfunction including abnormal pronation, insufficient shock absorption can occur, which may overload the lower extremities and the spine (Voloshin & Wosk, 1982). This reduced shock absorption through the kinetic chain has been associated with the presence of LBP (Voloshin & Wosk, 1982). As a result of this mechanical stress, lumbar intervertebral discs of the spine can undergo degenerative changes which impair their ability to appropriately transmit and dissipate these shock waves (Voloshin & Wosk, 1982). Alterations to these discs have been suggested to cause LBP symptoms in the majority of patients (Voloshin & Wosk, 1982). From a neuromuscular control perspective, the loading rate of the shock wave (the load amplitude divided by the time to reach the maximum amplitude) is an important consideration with respect to LBP mechanisms (Ogon, Spratt, Pope, & Salzman, 2000). Specifically, repetitive, rapidly applied impulsive loading has been shown to produce degenerative changes (Radine, Yang, Riegger, Kish, & O’Connor, 1991) which can lead to altered mechanics and cumulative damage to the spine. When shock absorption mechanisms are functioning properly, and the lumbar spine sufficiently stabilizes (in terms of magnitude and timing), the spine can appropriately respond to this shock wave (Ogon et al., 2000). However, when these mechanisms are impaired, the shock wave may act on the spine prior to its appropriate stabilization, and can therefore cause injury (Ogon et al., 2000). Thus, biomechanical dysfunction of the foot disrupts the normal mechanics of the kinetic chain which through different mechanisms, has been postulated to cause LBP as a result of cumulative damage.

While this pronation and impact-related pathology of the foot is an important consideration, especially within the gait literature, the role of proprioception is also of critical importance as predominantly evidenced within the balance literature. When considering the three main sensory systems which contribute to postural control (somatosensory, vestibular, and visual), and the role of the foot as the foundation of the body, the foot serves an important somatosensory function based on proprioceptive inputs (Horak & Macpherson, 1996; Losa Iglesias et al., 2012). Through proprioceptive feedback, the foot provides sensory information to the CNS through ascending pathways (Perry, Radtke, McIlroy, Fernie, & Maki, 2008). This information, along with vestibular and visual feedback, is integrated by the CNS, and then used by the musculoskeletal system to achieve appropriate postural orientation and balance during upright standing (Horak & Macpherson, 1996). Overall, based on these three mechanisms discussed, a variety of different insoles have been used to improve prevent and treat LBP through lumbopelvic and postural control as will be discussed below.

Insoles Intervention

Mechanical LBP, which accounts for 90% of all LBP (Cohen et al., 2008), is strongly suggested to be caused by biomechanical abnormalities (Kendall et al., 2014) which can often begin with the foot. Thus, many clinical practitioners frequently prescribe foot orthotics as an adjunct to other therapeutic treatments to address biomechanical dysfunctions of the foot in an attempt to normalize the kinetic chain (Cambron et al., 2011; Kendall et al., 2014; Michaud, 1997). When examining foot orthotics intervention studies, while the literature is equivocal, several studies have observed improvements in LBP associated with the use of these devices. While some of these studies used clinician assessments of LBP (Mattila et al., 2011; Milgrom, Finestone, Lubovsky, Zin, & Lahad, 2005), most of these studies formed their conclusions using self-report evaluations of LBP/disability based on questionnaires (Cambron et al., 2011; Castro-Méndez et al., 2013; Ferrari, 2007; Rothbart et al., 1995; Shabat et al., 2005) and pain scales such as the VAS (Cambron et al., 2011; Castro-Méndez et al., 2013) and the Quebec Back Pain Disability Scale (Dananberg & Guiliano, 1999).

A number of different types of foot orthotics are designed to prevent and treat a variety of musculoskeletal conditions. Thus, there are several different purposes these devices are meant to serve. These include increased shock absorption, relief to pressure-sensitive plantar areas to decrease pain, reduced plantar shearing forces, support of the foot in the position most desirable for weight bearing (neutral subtalar joint), the correction of functional deformities to shift the location of weight-bearing on the foot, and finally, increased plantar somatosensation (Lockard, 1988; Rothbart, 2005).

The prevention of abnormal foot pronation is one of the most commonly investigated biomechanical dysfunctions that foot orthotics are designed to address (Landorf, Keenan, &

Rushworth, 2001; Lockard et al., 1995; Rothbart et al., 1995). Foot orthotics dealing with this particular dysfunction are typically made of semi-rigid or rigid materials, and include specifically placed material reinforcements (posts) such as a medial wedge to control the pronation (Donatelli et al., 1988; Lockard, 1988). With this design, these types of foot orthotics work to position the subtalar joint in its neutral position at heel strike, during the stance phase of gait (Donatelli et al., 1988; Lockard 1988). By creating a more neutral position of the subtalar joint, these types of foot orthotics aim to improve the stability and alignment of the foot and its proximal structures (Bird & Payne, 1999; Donatelli et al., 1988; Lockard, 1988). By preventing abnormal foot pronation, these devices are suggested to normalize the kinematic chain by decreasing the internal rotation of the tibia and femur, anterior pelvic tilt, and lumbar lordosis which have been thought to be associated with LBP (Barwick et al., 2012; Cambron et al., 2011; Larsen et al., 2002). This foot to back dysfunction is largely based on theoretical models, as opposed to empirical biomechanical evidence. Thus, the mechanisms underlying the effects of foot orthotics are still not well understood.

Foot orthotics are often designed to structurally support and brace the foot in a position which is more optimal for weight-bearing. However, using a unique type of insole, improved foot function may alternatively be achieved through proprioceptive neuromuscular control mechanisms. Instead of bracing the foot with rigid reinforcements which may weaken the muscles in the foot, the insole used in the present study (6 Step Active 3/4 Length, Barefoot Science ^{TM}, Mississauga, Canada) is designed to allow the foot to experience a larger range of three-dimensional motion which is more reflective or barefoot motion, to stimulate and strengthen the intrinsic muscles of the foot and ultimately, the medial-longitudinal arch (Barefoot Science^{TM} 2007; Barefoot Science Products and Services Inc., 2016). Classified as a semi-rigid insole, these insoles consist of a fabric, light-weight three-quarter length material with a pronounced, rigid dome shaped arch located at the midfoot. The design of these insoles is quite unique compared to other insoles in that they use a multi-level foot strengthening system. This strengthening system involves a progression through six levels of foot-stimulating tabs (from smallest/softest to largest/firmest) which insert on the under-side of the insole into a cavity contained within a dome-shaped arch, and directly below the apex of the medial longitudinal arch (Figure 1). According to the manufacturers, these foot-stimulating tabs coupled with the dome-shaped arch are proposed to strengthen the intrinsic muscles of the foot through proprioceptive mechanisms by introducing a gentle recoil pressure to the medial longitudinal arch of the foot which leads to a reactive withdrawal reflex (Barefoot Science^{TM}, 2007; Barefoot Science Products and Services Inc., 2016). Through tactile and proprioceptive receptors in the feet (ie; muscle spindles), these insoles may modulate afferent neural pathways which alter the timing and magnitude of muscles activations locally in the foot, and more proximally up the kinetic chain, in turn, contributing to overall postural control (Barwick et al., 2012; Christovao et al., 2013; Dankerl et al., 2014). Over time, this mechanism is suggested to stimulate, strengthen and stabilize the foot through improved somatosensory feedback. The manufacturers make several claims about the effects of their insoles as a result of the proposed foot strengthening. Some of these include pain reduction of several body parts ranging from the feet to the shoulders, increased arch height, improved gait parameters, and immediate balance improvements (Barefoot Science Products and Services Inc., 2016). Considering these claims in the context of the LBP foot orthotics literature, these insoles may have implications in terms of LBP prevention and/or treatment. Thus, unlike many foot orthotics which are designed to improve foot function through a supportive, bracing mechanism, the insoles used in the current study use a proprioceptive neuromuscular control approach which requires further investigation.

Figure 1: Barefoot Science^{TM}, 6 Step Active 3/4 Length insoles. The inserted foot-stimulating tabs increase in size and firmness as they progress from level 1 (shown on the left, middle frame) to level 6 (middle frame on the right, and inserted into the insole in the bottom frame). The tab is inserted on the underside of the insole, beneath the midfoot, directly below the apex of the medial longitudinal arch.

This proprioceptive mechanism has predominantly been investigated within the insoles and balance literature. When reviewing the literature, although the results are mixed, many studies have observed a beneficial effect of insoles on balance as evidenced by a reduction in COP measures (Hamlyn et al., 2012; Losa Iglesias et al., 2012; Olmstead & Hertel, 2004; Palluel, Olivier, & Nougier, 2009; Qiu et al., 2012, 2013). The improvements that have been seen in the balance literature have been proposed to occur as a function of enhanced foot proprioception and the resulting mechanics (neuromuscular, kinematic, kinetic) which are transferred up the kinetic chain (Christovão et al., 2013).

Surface EMG (Murley, Landorf, & Menz, 2010; Murley & Bird, 2006; Tomaro & Burdett, 1993) and kinematic (McClay & Manal, 1998; McPoil & Cornwall, 2000; Nigg et al., 1993) measures, while routinely examined in the insole gait literature, are rarely used as outcome measures when balance tasks are investigated. In addition, within this body of literature, these techniques have generally been used to assess alterations in the mechanics of the lower extremities. While the lower extremity has been theoretically linked to the function of the lumbopelvic region, a direct examination of the biomechanical effects of insoles on this more proximal region is lacking within the literature. When the lumbopelvic and thigh regions have been examined, EMG findings are mixed, with significantly changed (Hertel et al., 2005; Mündermann et al., 2006; Nawoczenski & Ludewig, 1999) and unchanged (Bird et al., 2003; Dingenen et al., 2015) muscle activations as a result of insole (or heel lifts/foot wedges) use. The effects of insoles on kinematic outcomes for the lumbopelvic region are also mixed, with significant changes that were minimal (Marinakis & Catalfamo, 2004; Nester et al., 2003) or no changes at all (Esfandiari et al., 2013).

Within the context of the present study, the lumbopelvic region is of critical importance due to its central role in postural control and its strong association with LBP. Therefore, the current study sought to provide a comprehensive, and much needed assessment of the effect that insoles (described previously) may have on lumbopelvic control through the use of EMG, kinematic, and kinetic measures. As insoles have been suggested to improve balance through improved proprioceptive feedback, the present study targeted this system through the use of a relative change analysis and balance tests that challenge postural control by manipulating the somatosensory information available during these tests.

2.3.5 Functional Impairments, Clinical Tests, and Interventions of Lumbopelvic Control and Balance Associated with Low Back Pain Summary

As evidenced in this review, optimal motor control heavily depends on the intact proprioception and stability of the lumbar spine. This is particularly evident in the context of LBP, such that impairments in proprioception and spinal stability have been suggested to be associated with motor control impairments in individuals with LBP. Strong evidence of this has been cited within the lumbopelvic control and balance literature during functional tasks and clinical tests. Due to the postulated association between the functioning of the foot and lumbopelvic region, and with its implications for postural control, insoles which use a proprioceptive mechanism may have beneficial effects in the proximal lumbopelvic region and may improve postural control. Using a postural context which particularly challenges the somatosensory system (eyes closed condition and balance tests with a narrow/small BOS), this proprioceptive insole mechanism can be indirectly targeted through the use of a relative change analysis which was used within the present study.

CHAPTER 3

Introduction

3. Introduction

The foot has been suggested to have major, but largely theoretical, implications for lumbopelvic and postural control within a LBP context (Barwick et al., 2012; Bird & Payne, 1999; Hamlyn et al., 2012). It is therefore, not surprising that insoles are commonly prescribed as an intervention for LBP (Kendall et al., 2014). There is little empirical and objective biomechanical evidence however, as to the mechanism behind the influence insoles may have on LBP which has mostly been based on self-report data (Cambron et al., 2011; Shabat et al., 2005). In addition, the lumbopelvic region, in spite of being implicated in LBP, has been largely under investigated in the insoles literature, but when tested, has produced minimal and equivocal results (Bird et al., 2003; Esfandiari et al., 2013; Marinakis & Catalfamo 2004; Nester et al., 2003).

Lumbopelvic control is a critical component of balance (Saunders et al., 2005), and literature that has investigated the influence that insoles have on balance has provided a potential mechanism through which they may reduce LBP. Improvements in balance therefore, may be reflective of improved lumbopelvic control. These studies suggest that insoles enhance the somatosensory (tactile and proprioceptive) information on the plantar surface of the foot (Hamlyn et al., 2012; Hijmans et al., 2007), which in turn informs changes in neuromuscular control that leads to improvements in balance and posture during standing (Hamlyn et al., 2012; Hijmans et al., 2007; Mochizuki & Amadio, 2003; Olmstead & Hertel, 2004; Rothbart, 2005). There is a lack of literature however, that investigates the effects of insoles on lumbopelvic control using balance tests and biomechanical measures such as EMG and kinematic outcomes.

Therefore, the first aim of this study involved the use of an insole (Barefoot Science Products and Services Inc., 2016), specifically designed to influence the somatosensory system, as a means of investigating the effects of this insole on lumbopelvic control and balance. The influence of insoles on lumbopelvic control was investigated using the sidelying AHAbd test which has been previously shown to have moderately strong associations with LBP (Babiolakis et al., 2015; Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2009). In addition, by using an eyes closed condition which targets the proprioceptive system by removing visual input, and standing balances tests with small and/or narrow bases of support, this study aimed to investigate the hypothesis that changes in the somatosensory system drive the influence of insoles on balance outcomes.

The AHAbd test is of particular interest as a lumbopelvic control test as it was the *only* test out of several to distinguish between PDs and NPDs during prolonged standing (Nelson-Wong & Callaghan, 2010a; Nelson-Wong et al., 2009) and between nurses with and without a recent history of LBP due to back injury (Babiolakis et al., 2015). One drawback to the sidelying AHAbd test however, is that it is does not use a functionally relevant posture such as standing. As such, the second aim of this study then was to determine if standing balance tests were associated with the AHAbd test, to be used as more functionally relevant clinical markers of transient LBP development during standing.

CHAPTER 4

Methods

4. Methods

4.1 Participants

Twenty (9 males and 11 females), unaffected or healthy participants were recruited for this study from the university and general community populations, mean age 23.40 yr ±3.62, by word of mouth and through e-mail communications (Table 1). The term unaffected as it applies to the participants within this study reflects a non-clinical population and is specifically defined by the inclusion criteria outlined below. Inclusion criteria included no current use of prescribed orthotics; no prior history of pain or injury in the back, legs, or feet that required medical treatment and/or resulted in more than three days off of school or work; no previous back, hip, leg, or foot surgery; the ability to stand for more than four hours and to walk for more than sixty minutes. All participants provided written informed consent prior to participation in the study and all procedures were approved by the York University Human Participants Review Committee. Data were collected as part of a larger study investigating the effects of insoles on EMG, kinematic, and kinetic measures of the trunk and lower extremities during various tasks including gait, lifting, and standing.

Table 1: Participant descriptive characteristics. Mean (SD) or n (%) are reported. Note: BMI: Body Mass Index.

Characteristics	Female (n=11)	Male (n=9)	Total (n=20)
Age (yrs)	23.09 (4.21)	23.78 (2.95)	23.40 (3.62)
Height (m)	1.64 (0.07)	1.83 (0.07)	1.73 (0.11)
Body Mass (kg)	62.72 (8.09)	81.79 (6.21)	71.30 (12.06)
BMI (kg/m^2)	23.15 (2.28)	24.58 (2.02)	23.80 (2.23)
Leg Dominance n (%)			
Right	10 (90.91%)	8 (88.89%)	18 (90.00%)
Left	1 (9.09%)	1 (11.11%)	2 (10.00%)

4.2 Laboratory preparation

Seven motion capture cameras (Vicon MX, Vicon Systems Ltd, Oxford, UK) were positioned and focused around the collection space and a standard dynamic calibration of the motion capture system was performed (Woolard, 1999). To set the global (laboratory) coordinate system origin, a calibration wand was placed with the identical position and orientation in the centre of the collection space (marked by an X) every data collection. Subsequently, the force platform amplifier (AMTI-OR6, AMTI, Massachusetts, USA) was zeroed to remove any bias.

4.3 Instrumentation

Surface EMG was recorded using three amplifier systems (AMT-8, Bortec Biomedical Ltd., Calgary, Canada) and pairs of silver/silver-chloride electrodes (Ambu® Blue Sensor N, Ambu A/S, Denmark) with a centre-to-centre spacing of approximately 2.5 cm. These pairs of electrodes were applied over the muscle bellies of twelve bilateral trunk, hip, and thigh, and shank muscles, only eight of which were of interest for this thesis: the thoracic erector spinae (TES); lumbar erector spinae (LES); RA; EO; IO; GTMED; BF; and VM muscles (Table 2,

Figure 2). Three ground (reference) electrodes, one for each amplifier system, were placed on the R/L clavicles and on the left tibial tubercle. The raw EMG signals were differentially amplified (frequency response 10 Hz – 1000 Hz common-mode rejection ratio 115 dB at 60 Hz, input impedance 10 GΩ) and converted at 2400 Hz from analog to digital.

Table 2: Anatomical descriptions of muscle electrode placement for the eight bilateral trunk, hip, and thigh muscles examined.

Body Region	Muscles	Electrode Placements	References
Trunk	Thoracic erector spinae (TES)	~5 cm lateral to T_9 spinous process, over the largest muscle mass	Drake, Fischer, Brown & Callaghan, 2006
	Lumbar erector spinae (LES)	~3 cm lateral to L_3 spinous process, over the largest muscle mass	Drake et al., 2006
	Rectus abdominis (RA)	~3 cm lateral to the umbilicus	Drake et al., 2006
	External obliques (EOs)	~15 cm lateral to the umbilicus	Drake et al., 2006
	Internal obliques (IOs)	Below the EO electrodes and just superior to the inguinal ligament	Drake et al., 2006
Hip	Gluteus medius (GTMED)	~2.54 cm distal to the midpoint of the iliac crest	Nelson-Wong & Callaghan, 2010a
Thigh	Biceps femoris (BF)	~50% on the line between the ischial tuberosity and the lateral epicondyle of the tibia	Hermens, Freriks, Disselhorst-Klug, & Rau, 2000
	Vastus medialis (VM)	~80% on the line between the anterior superior iliac spine and the joint space in front of the anterior border of the medial ligament of the knee	Hermens et al., 2000

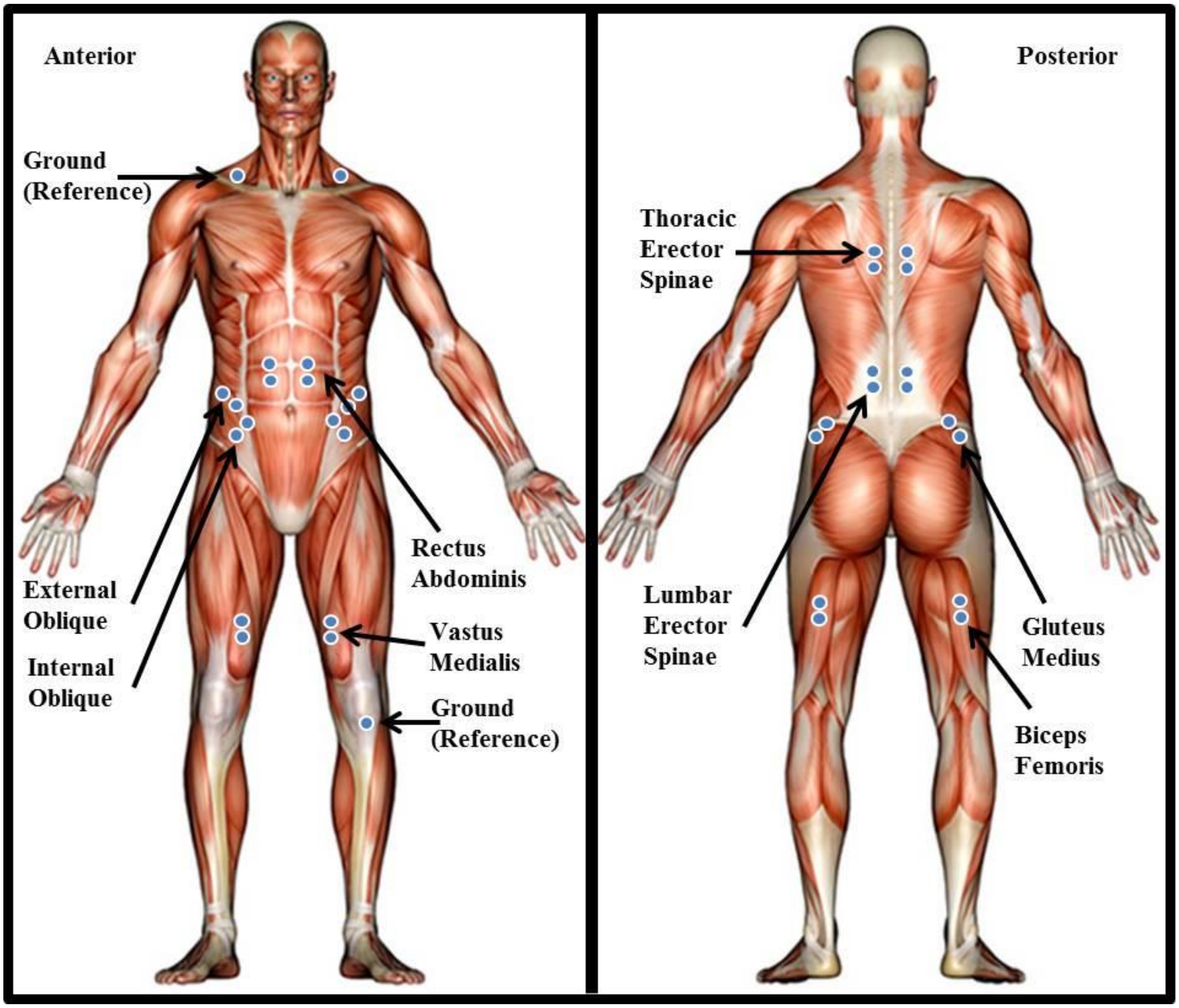

Figure 2: Representation of EMG electrode placement. Anteriorly there were three ground electrodes and eight pairs of electrodes placed over four bilateral muscles. Posteriorly, there were eight pairs of electrodes placed over four bilateral muscles. In total, 16 muscles (8 bilaterally) were examined (Retrieved and adapted from http://www.paulabrown.net/human-body-diagram-blank).

Three-dimensional full body kinematics were recorded at 50 Hz using a motion capture system. Seventy-five passive reflective markers, 21 of which were of interest for the present study, were adhered to each participant. Two clusters (5 markers each) were affixed over the T_{12} and L_5 vertebral levels (Figure 3). The remaining markers were placed on the trunk (C_7 vertebral level, and R/L acromia for a total of 3 markers), pelvis (R/L anterior superior iliac spines; R/L posterior superior iliac spines; and R/L iliac crests for a total of 6 markers); and on the R/L greater trochanters (2 markers) (van Sint Jan, 2007, Figure 4). Three-dimensional ground reaction forces and moments were recorded at 2400 Hz using a single force platform.

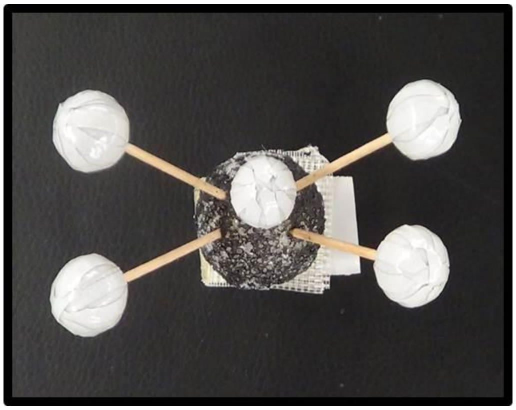

Figure 3: Representation of a passive reflective marker spine cluster. The spine clusters were placed at the T_{12} and L_5 spinal levels to provide data that was used to define three rigid segments (trunk, lower thoracic spine, and lower lumbar spine).

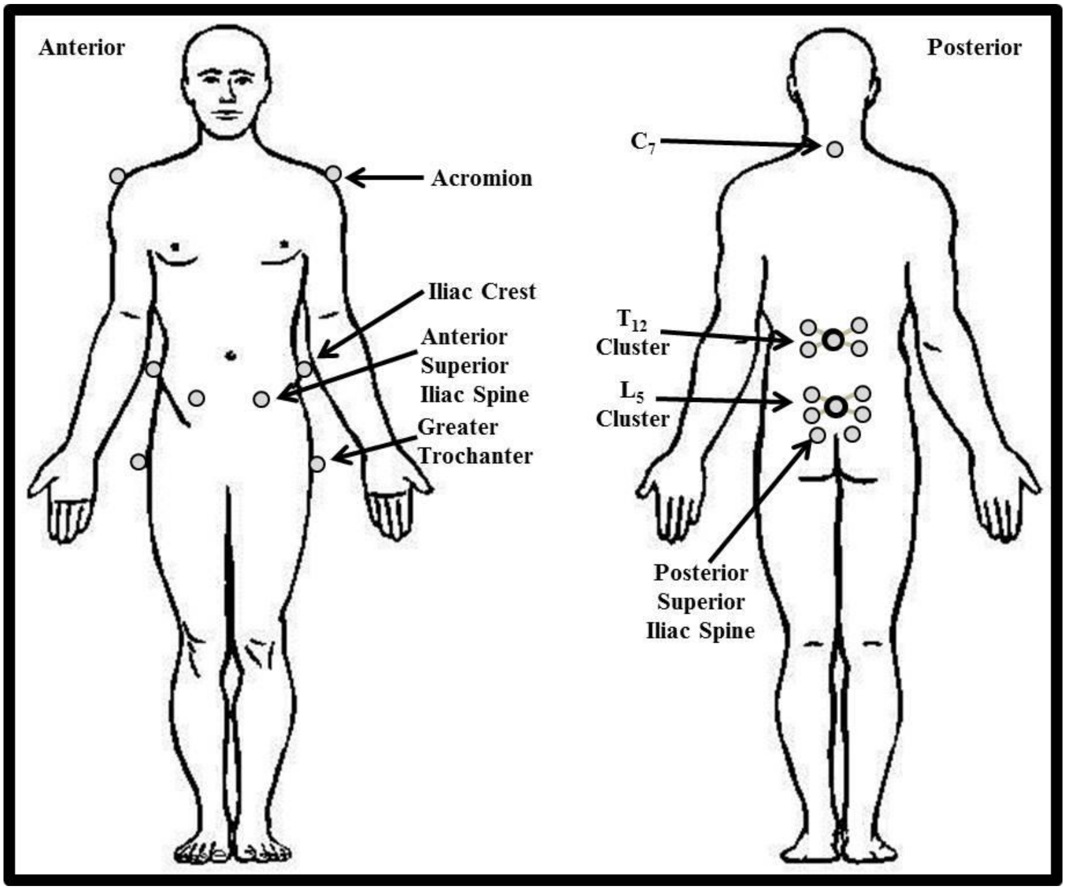

Figure 4: Representation of the passive reflective marker placement. These markers were used to define four rigid segments (trunk, lower thoracic spine, lower lumbar spine, and the pelvis) which were used in the calculation of two global (trunk and pelvis) and two relative (lumbar and lumbopelvic) angles (Retrieved and adapted from http://medicalanatomy.net/outline-human-body/).

4.4 Procedures

This study was an intervention-based design consisting of two visits. At the first visit, each participant received a pair of insoles based on their shoe size. Following an average of 8.37 weeks of insoles use, participants returned and completed the second visit. The same protocol was used for visit one and two, with the only difference being that the participants wore the insoles during the visit two protocol.

Participants were asked to wear the insoles at all times when they were wearing shoes that were able to accommodate them. In line with the manufacturer’s instructions, participants were instructed to progress through the six level strengthening system whereby foot-stimulating tabs (ranging from smallest/softest to largest/firmest) were inserted on the under-side of the insole (in the cavity of the dome-shaped arch). Participants were expected to transition between levels once they no longer felt pressure from the arch (~ 4–7 days). If participants felt any extreme or prolonged (i.e. longer than 2 weeks) discomfort or irritation at a particular level, they were instructed to return to a lower level or to discontinue use.

Each visit consisted of seven blocks, beginning with (1) initial measures, which included the AHAbd test, and (2) an instrumented-walkway (force platform and pressure mat) shod and barefoot walking trials. These were followed by four major blocks: (3) balance (OLST and SRT), (4) standing, (5) lifting, and (6) treadmill walking trials. The final block (7) was a repeat of block two. The repeated trials in the balance block (3) were randomized to minimize the risk of order effects. For each participant, blocks one, two, and seven were always presented in the same order, and the four major blocks (3–6) were presented in a random order.

A series of initial measures (block one) were first collected. To gain an understanding of participants’ physical activity and sedentary exposures, they were asked to report how much time they spent in the following activities on average in the previous week: sitting, standing, walking, performing activities of daily living, and performing resistance and aerobic exercise. Participants were then asked to report their leg dominance based on which leg they use to kick a ball (Hoffman, Schrader, Applegate, & Koceja, 1998; Nelson-Wong et al., 2012). Anthropometric measures (height, body mass, and trunk depth) were taken to provide the necessary inputs for the construction of a custom three-dimensional kinematic model using Visual3D (v.4, C-Motion, Inc., Germantown, USA).

Participants subsequently underwent skin preparation (shaving and swabbing with rubbing alcohol) prior to the application of EMG electrodes on selected trunk, hip, and thigh muscles (mentioned previously in Section 4.3). This skin preparation was performed to improve the adherence of the electrodes to the skin. To allow for the normalization of EMG recordings, participants performed a series of isometric MVC protocols against manual resistance applied by the researchers as described in Table 3. Participants performed two trials of each MVC protocol for 3-5 seconds, with three minutes of rest between trials to minimize the effects of fatigue.

Table 3: Description of the protocols used for determining the maximal voluntary contractions (MVCs) for the eight bilateral trunk, hip, and thigh muscles examined.

Body Region	Muscles	MVC Protocol
Trunk	Thoracic erector spinae (TES) Lumbar erector spinae (LES)	The participant lay prone on a table, with their upper body hanging off the edge (at the anterior superior iliac spine level). Subsequently, they folded their arms across their chest, and performed a resisted trunk extension (McGill, 1992).
	Rectus abdominis (RA) External oblique (EO) Internal oblique (IO)	Using a modified sit-up protocol, the participant performed resisted trunk flexion, lateral bend, and axial twist movements (McGill, 1992).
Hip	Gluteus medius (GTMED)	The participant assumed a sidelying position and performed a resisted hip abduction (Nelson-Wong et al., 2008).
Thigh	Biceps femoris (BF)	In a prone position, the participant performed a resisted knee flexion with their knee flexed at 55° (Rutherford, Hubley-Kozey, & Stanish 2011).
	Vastus medialis (VM)	In a prone position, the participant performed a resisted knee extension, with the knee flexed at 15° (Rutherford et al., 2011).

To perform the AHAbd test, participants were instructed to assume a side lying position on the table, with both limbs extended and aligned with their trunk, and pelvis aligned in the frontal plane, perpendicular to the support surface (Nelson-Wong et al., 2009). All participants received standardized instructions: “Please keep your knee straight and raise your top thigh and leg towards the ceiling, keeping them in line with your body, and try not to let your pelvis tip forwards or backwards” (Nelson-Wong et al., 2009; Figure 5). Participants performed this test twice bilaterally for a total of four trials with EMG recorded. Prior to the EMG recording of the test, a researcher (C.S.B whenever possible), took each participant through the passive range of motion required for the AHAbd test. This was done to familiarize each participant with the movement sequence and range of motion for this test (Davis et al., 2011) and to ensure that the instrumentation cables were not interfering with the participants’ motion. The AHAbd test was rated on a 4-point ordinal scale (0–3) ranging from no loss of pelvis frontal plane position (smoothly, easily performs movement; lower extremities, pelvis, trunk and shoulders remaining aligned in the frontal plane) to a severe loss of pelvis frontal plane position (More than 3 of the following characteristics: noticeable wobble through movement; tipping of the pelvis, trunk, or shoulder rotation; increased hip flexion and/or rotation of the moving limb; and rapid or uncontrolled movement, and/or unable to regain control of movement once lost or may lose balance and may have to place hand on the table – Davis et al., 2011; Nelson-Wong et al., 2009). To verify scores post-collection and to allow for the quantification of inter-rater and intra-rater reliability (see Section 4.5.4), the AHAbd test was videotaped from the sagittal view.

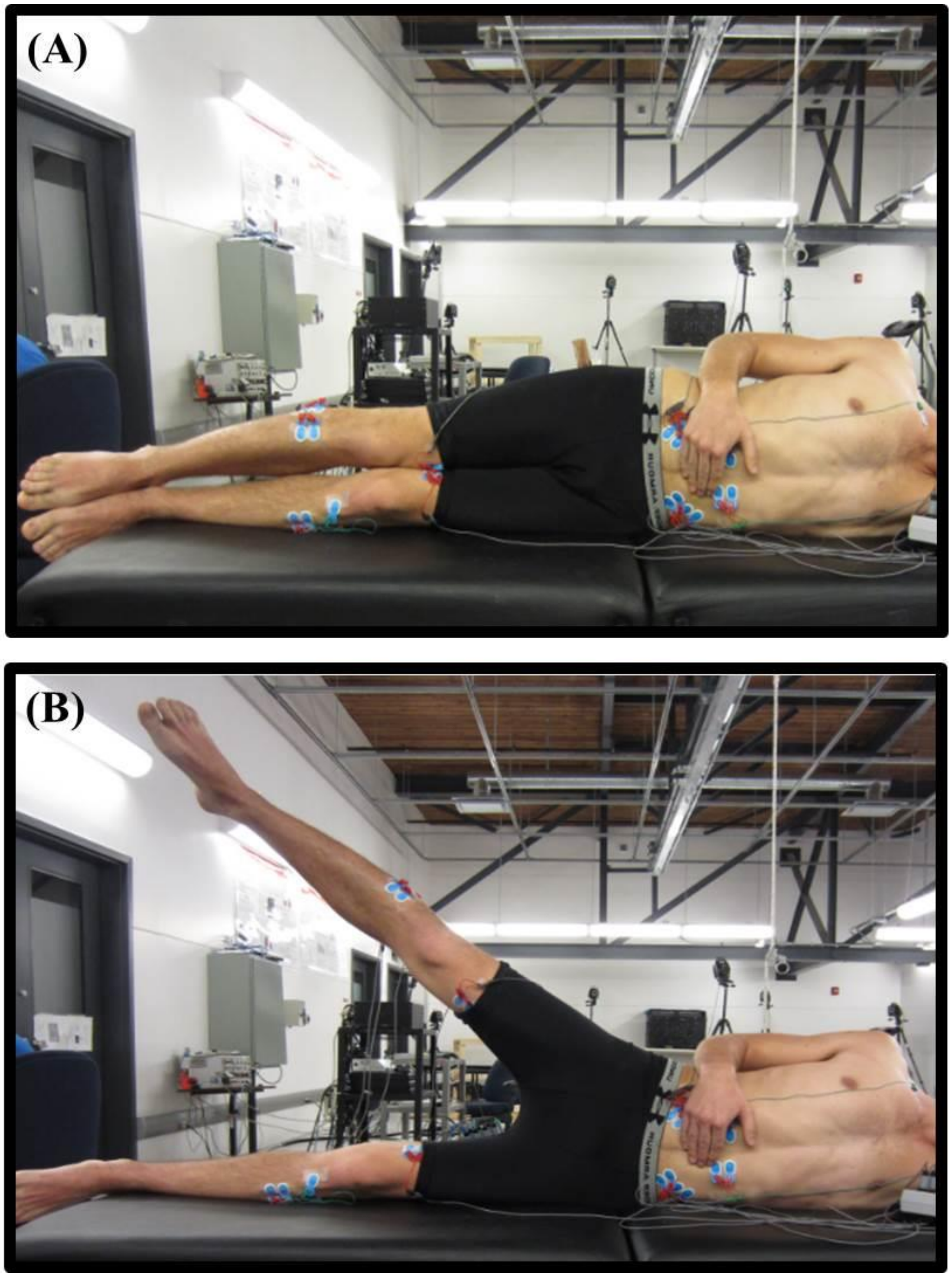

Figure 5: Representation of the starting (top panel) and test (bottom panel) positions of the active hip abduction (AHAbd) test. Participants were instructed to raise their top thigh towards the ceiling without letting their pelvis tip forwards or backwards. The AHAbd test was used to determine the level of lumbopelvic control in the frontal plane.

Following the AHAbd test, passive-reflective markers were applied to the participants to capture three-dimensional kinematics. A static calibration trial was subsequently collected for the set-up of a custom kinematic model (see Section 4.5.2). This involved the participant standing quietly with their head in a neutral position, their feet pointing forward, and their arms abducted to 90° in a ‘T-pose’. A 10 second upright standing trial was also collected and used to zero the calculated angles during kinematic processing (see Section 4.5.2). For the upright standing trial, participants stood quietly with their head in a neutral position, their feet pointing forward, and their arms loose, in a relaxed position by their sides.

For block three (balance), EMG, kinematic, kinetic, and timing measures were recorded as participants performed the OLST and SRT on a single force platform (Figure 6). Participants performed both tests shod, with their eyes open and closed. Two repeats of each condition were completed in a random order, for a total of eight OLST trials (2 eyes open right stance, 2 eyes open left stance and 2 eyes closed right stance, 2 eyes closed left stance) and four SRT trials (two eyes open and two eyes closed). Participants were provided verbal instructions, a demonstration of specific postures, and the termination criteria associated with each test. For the OLST, participants’ feet were placed in a standardized position of 0.17 m stance width (McIlroy & Maki, 1997) marked by two pieces of tape on the force platform. The OLST required participants to initially stand in a relaxed position, with their weight evenly distributed between both feet (Jonsson et al., 2004). Participants then attempted to maintain their balance while standing freely on one leg for a maximum time of 30 seconds (Briggs et al., 1989; Jonsson et al., 2004). The time was started when the subject assumed the proper position and ended when participants disengaged from the single stance position by displacing the foot they were standing on from the original position, by using the foot of their swing leg to support their stance leg, or by touching the support surface with the foot of their swing leg (Briggs et al., 1989; Rogers et al., 2003). The test was also ended if participants opened their eyes during eyes closed trials, or if they reached the maximum balance time of 30 seconds (Briggs et al., 1989; Rogers et al., 2003). Post-collection, a measure of pelvic drop was assessed for the OLST trials using the custom kinematic model created for each participant in Visual3D (see Section 4.5.4).

For the SRT, each participant stood in a tandem heel-to-toe stance with their arms folded across their chest (Johnson et al., 2005; Newton, 1989) for a maximum time of 30 seconds (Jonsson et al., 2005; Morioka et al., 2000). Timing began once the participants assumed the proper position and ended when participants moved their feet from the original starting position, unfolded their arms, opened their eyes during eyes closed trials, and/or if they reached the maximum balance time of 30 seconds (Johnson et al., 2005; Newton, 1989). To fit both of their feet on the force platform during the SRT, participants were required to position their feet diagonally, from the back right to the front left corner of the force platform. The participants self-selected whether their dominant or non-dominant foot was behind and they were instructed to keep this consistent for all of the SRT trials.

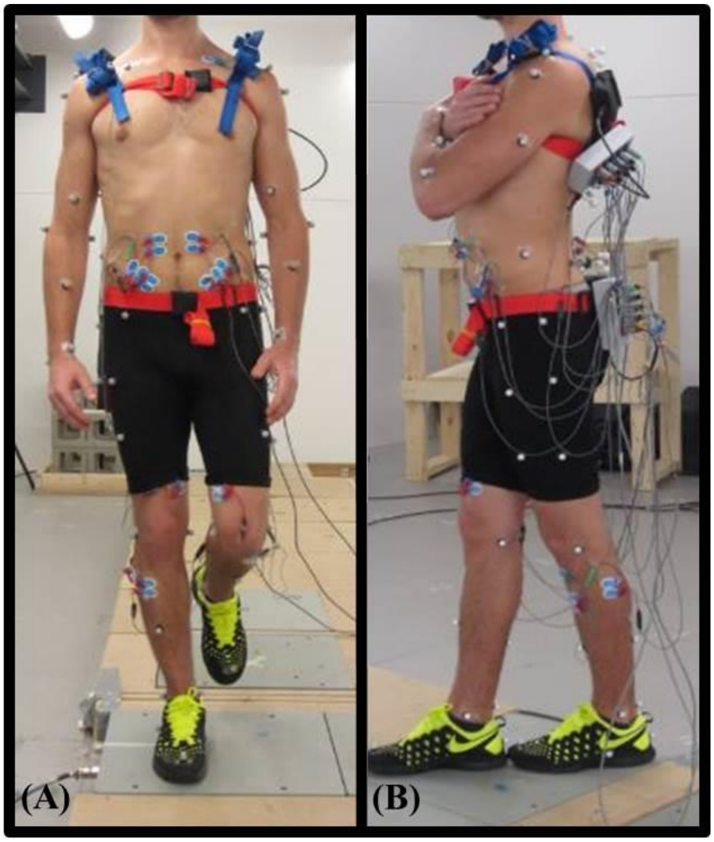

Figure 6: A representation of the two standing balance tests. **A)** One-leg stance test (OLST) performed on the right leg. Each participant, initially stood with their weight evenly distributed between both feet, were tasked with shifting towards, and maintaining their balance over, either the right or left leg for a maximum of 30 seconds. **B)** Sharpened Romberg test (SRT). Each participant stood in a tandem stance (heel-to-toe), with their arms across their chest for a maximum of 30 seconds.

During block four (standing), participants completed a 30 minute occupational standing simulation (Gallagher et al., 2011; Gregory et al., 2008; Nelson-Wong & Callaghan 2010a; Nelson-Wong et al., 2008). For the current study, the occupational simulation involved card dealing to mimic a casino dealer (Gregory et al., 2008). Participants were instructed to stand comfortably with one foot on each force platform while performing the card dealing task using a non-adjustable work table positioned directly in front of them. Throughout the entire protocol, participants were not permitted to lean their body weight (upper and lower extremities) on the work table and participants had to keep each of their feet on their respective force platforms. A VAS was administered to assess participants' transient LBP development during standing. It was presented on a piece of paper as a 100 mm non-hatched line with anchors at the left, 'No pain', and right, 'Worst pain imaginable'. Every 15 minutes (at the beginning, middle, and end of standing) participants marked their current level of LBP. Each VAS was presented separately, not allowing participants to refer to their previous VAS markings as this could have skewed their scores (Kelly, 1998). Post-collection, the VAS was quantified by measuring the distance (mm) between the left side of the scale ('No pain') and the vertical line marked by the participant.

4.5 Data Processing for the Insoles

All EMG, kinematic, and kinetic signal processing was performed using Visual3D. To compare between the eyes closed and eyes open conditions for the balance tests (OLST and SRT), EMG, kinematic (only the OLST), kinetic, and balance time outcomes were quantified using the relative change equation (Equation 1) presented in Section 2.3.3. The interpretation of this equation is detailed in Section 5.

4.5.1 Electromyography

Raw EMG signals were high-pass filtered using a dual-pass, 4th-order Butterworth filter with a cut-off frequency of 30 Hz, to remove heart rate contamination (Drake & Callaghan, 2006). Signals were then full-wave rectified and low-pass filtered using a dual-pass, 4th-order Butterworth filter with a cut-off frequency of 2.5 Hz (Brereton & McGill, 1998). The maximum value from the two MVC trials for each recorded muscle was selected as the MVC. This value was used to normalize the EMG data to %MVC, to be able to compare between participants. EMG data were down sampled from 2400 Hz to 50 Hz as a data reduction measure and to line up the EMG and kinematic data in time. Maximum (AHAbd test, OLST, and SRT) and mean

(OLST and SRT) activation levels were determined for each muscle. These values were then averaged across repeated trials for each participant, and relative change values were calculated for the OLST and SRT (Equation 1).

4.5.2 Kinematics

Using the static calibration (‘T-pose’) and anthropometric measurements (height, body mass, and trunk depth) from the data collection, a custom three-dimensional model was constructed for each participant in Visual3D. The marker data was used to create four rigid segments: the trunk (defined by the acromia, iliac crests, C_7, and T_{12} and L_5 spine clusters), lower thoracic spine (defined by the T_{12} spine cluster), lower lumbar spine (defined by the L_5 spine cluster), and the pelvis (defined by the iliac crests, greater trochanters, and the anterior and posterior superior iliac spines) (C-Motion Research Biomechanics Wiki-Documentation Marker Set Guidelines, http://c-motion.com/v3dwiki/index .php?title=Marker_Set_Guidelines). Two markers at the proximal and distal ends of the segment were used to calculate the segment endpoints (Table 4), and the remaining markers were used to track the segment. The segment endpoints were calculated as the mid-point of the distance between the two markers at the proximal and distal ends. Segment coordinate systems were then computed based on these markers and endpoints. Further details outlining how the marker data were used to define these segments and how the segment coordinate systems were computed are included in the Appendix.

Table 4: Passive reflective markers used to define the segment endpoints (the mid-point of the distance between the two markers at the proximal and distal ends).

Segment	Proximal Endpoint Markers	Distal Endpoint Markers
Trunk	Iliac crests (R/L)	Acromia (R/L)
Lower Thoracic Spine (T_{12}) and Lower Lumbar Spine (L_5)	Two inferior markers of the respective cluster	Two superior markers of the respective cluster
Pelvis	Iliac crests (R/L)	Greater trochanters (R/L)

Segment coordinate systems were consistent with the global (laboratory) coordinate system which was defined based on the position and orientation of the calibration wand in the capture space during the laboratory preparation. The X, Y, and, Z axes were defined as the M-L, A-P, and supero-inferior axes, respectively. Relative to the participant, the positive directions for each axis were: M-L (Y) axis (right); A-P (X) axis (forward); supero-inferior (Z) axis (upward) (Figure 7). Based on the orientation of these axes, a Cardan X-Y-Z rotation sequence was used to calculate all angles. With this rotation sequence, rotation around the X, Y, and Z axes represented flexion-extension, lateral bend, and axial twist, respectively. To interpret the angles, positive/negative values were defined as flexion/extension (around the X axis), right/left lateral bend (around the Y axis), and right/left axial twist (around the Z axis).

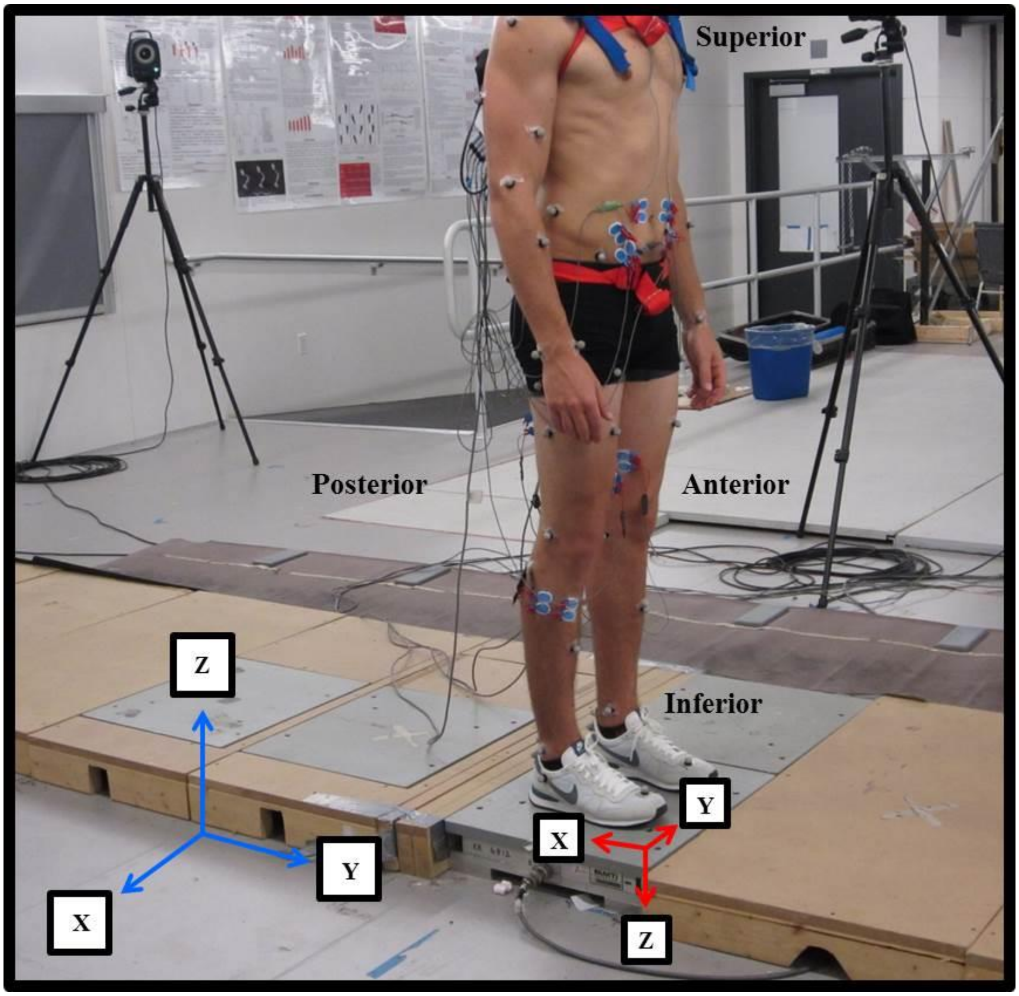

Figure 7: A representation of the global coordinate system of the laboratory (blue arrows) and the force platform coordinate system (red arrows) relative to the participant. Kinematic measures were based on the global coordinate system in which positive/negative values were defined as flexion/extension (around the X axis), right/left lateral bend (around the Y axis), and right/left axial twist (around the Z axis). Kinetic measures were based on the force platform coordinate system in which positive/negative centre of pressure (COP) values were defined as backward/forward (along the antero-posterior (A-P) X axis), left/right (along the medio-lateral (M-L) Y axis) excursions along the Y-axis, and downward/upward (along the supero-inferior Z axis).

Two global (trunk and pelvis) and two relative (lumbar and lumbopelvic) angles were then calculated. All global angles were calculated with respect to the global (laboratory) coordinate system, and all relative angles were calculated using two segments of interest (lumbopelvic: T_{12} with respect to the pelvis, and lumbar: T_{12} with respect to L_5). Angles were then low-pass filtered using a dual-pass, 4th-order Butterworth filter with a cut-off frequency of 6 Hz (Winter, 2009). To zero all of the angles, the mean value for each angle from the upright standing trial was subtracted from the angles calculated for the OLST trials. This was done to account for intra-individual differences in the starting postures of participants. Maximum and mean angles were determined for each region (lumbar, pelvis, lumbopelvic, and trunk) in all three axes separately for the OLST trials. These values were then averaged across repeated trials for each participant, and relative change values were calculated (Equation 1). To discern which direction of motion produced the greatest values, maximums were determined by selecting the highest absolute value. For instance, with regard to the flexion-extension (X) axis, if the magnitude of a negative value (indicating extension) was higher than the magnitude of a positive value (indicating flexion), the negative value was selected as the maximum.

In addition to analyzing participant’s motion in each axis separately, a Euclidean norm angle was calculated as a means of quantifying overall three-dimensional motion using the following Equation 2 (Graham & Brown, 2012):

Equation 2: $r\,(t) = \sqrt{x(t)^2 + y(t)^2 + z(t)^2}$

The Euclidean norm angle was calculated for the time-varying signal $r\,(t)$ by summing the squares of each data point for the flexion-extension (X), lateral bend (Y), and axial twist (Z) axes and taking the square root of this value at each data point in time for the entire signal. Maximum and mean Euclidean norm angles were determined for each region (lumbar, pelvis, lumbopelvic, and trunk). These values were then averaged across repeated trials for each participant, and relative change values were calculated (Equation 1).

4.5.3 Kinetics

Kinetic data were down sampled from 2400 Hz to 50 Hz as a data reduction measure and to line up the kinetic and kinematic data in time. For the OLST and SRT trials, the time-varying COP_d signals were calculated for the A-P (X) and M-L (Y) axes by dividing the moments about the Y (M_y) and X (M_x) axes, respectively, by the ground reaction force along the Z axis (F_z) using the following Equations 3 and 4 (Hewson, Singh, Snoussi, & Duchene, 2010):

Equation 3: A-P $COP_d = -M_y/F_z$

Equation 4: M-L $COP_d = M_x/F_z$

These calculations were performed based on the force platform coordinate system relative to the participant, such that the X, Y, and Z axes represented the A-P, M-L, and supero-inferior axes, respectively. The positive directions of the three axes were defined as follows: A-P (X) axis (backward); M-L (Y) axis (left); supero-inferior (Z) axis (downward) (Figure 7). The COP_d signals were zeroed for each trial by subtracting the mean value to account for the intra-individual bias introduced by the different testing postures that participants assumed each trial.

In addition to displacement, the root mean square (COP_{RMS}), total path ($COP_{totalpath}$), and velocity (COP_v) were also computed using the following Equations 5 (Winter, 2009),6 (Donath, Rother, Zahner, & Faude, 2012), and 7 (Ball & Best, 2007):

Equation 5: $COP_{RMS}\ s(t) = \sqrt{\frac{1}{N}\sum_{i=1}^{N} s(i)^2}$

Equation 6: $COP_{totalpath} = \sum_{i=2}^{n} \sqrt{(x_i - x_{i-1})^2 + (y_i - y_{i-1})^2}$

Equation 7: $$\mathrm{COP}_{\mathrm{v}\,n} = \frac{\mathrm{COP}_{\mathrm{d}\,(n+1)} - \mathrm{COP}_{\mathrm{d}\,(n-1)}}{2 * t}$$

The $\mathrm{COP}_{\mathrm{RMS}}$ represents the SD of the $\mathrm{COP}_{\mathrm{d}}$ (Palmieri et al., 2002). This was calculated for the time-varying signal s (t) by squaring each data point, calculating the mean of all of these squared data points (i = 1 to N = total number of data points), and finally, by taking the square root of this mean value. The $\mathrm{COP}_{\mathrm{totalpath}}$ is defined as the total distance travelled by the COP over the course of the trial duration (Palmieri et al., 2002) This was calculated by first taking the difference between each consecutive data point (i and i–1) and squaring this difference for both the A-P (X) and M-L (Y) axes. The square root of the sum of these squared differences was then taken, and finally, the sum of this resultant value for all of the data points (i = 2 to n = the total number of data points) was calculated. The $\mathrm{COP}_{\mathrm{v}}$ represents the first derivative of the time-varying $\mathrm{COP}_{\mathrm{d}}$ data. This was calculated by using the three-point central differences method. For each data point, the difference between the point ahead (n + 1) and behind (n – 1) the point of interest (n = sample at which the velocity is being calculated) was taken and then divided by two times the time interval (0.02s) between data points.

The $\mathrm{COP}_{\mathrm{d}}$, $\mathrm{COP}_{\mathrm{RMS}}$, and $\mathrm{COP}_{\mathrm{v}}$ were all calculated in both the A-P (X) and M-L (Y) axes separately, and the $\mathrm{COP}_{\mathrm{totalpath}}$ was calculated based on the resultant of these axes for the OLST and SRT trials. Absolute mean ($\mathrm{COP}_{\mathrm{d}}$ and $\mathrm{COP}_{\mathrm{v}}$) and peak-to-peak (maximum – minimum; $\mathrm{COP}_{\mathrm{d}}$) amplitude values were determined. The calculation of metrics was not applicable for the $\mathrm{COP}_{\mathrm{RMS}}$ and $\mathrm{COP}_{\mathrm{totalpath}}$. All of these values were then averaged across repeated trials for each participant, and relative change values were calculated (Equation 1). COP measures were calculated in millimetres and lower values (compared to higher values) were considered reflective of better postural control for this young, unaffected population (Palmieri et al., 2002).

4.5.4 Test Performance Outcomes and Perceived Low Back Pain

To establish inter-rater and intra-rater reliability, and the final scores for the AHAbd test in the present study, C.S.B (who has been trained in the administration and scoring of this test since 2012) and two graduate students rated this test. To train these graduate students, C.S.B led a video-based tutorial (provided by Dr. Erika Nelson-Wong, one of the researchers who developed this test), and reviewed manuscripts which detail the development and scoring of the test (Davis et al., 2011; Nelson-Wong et al., 2009). For inter-rater reliability, the three students rated the AHAbd test videos for all of the participants separately. After approximately three weeks (a ‘washout’ period to avoid the raters’ recollection of the videos and their previously assigned scores), the three students re-rated all of the videos in order to establish intra-rater reliability (Davis et al., 2011). To determine the final scores for the all of the AHAbd test videos, a consensus score based on the ratings of the three researchers was assigned to each. When there was a disagreement about a score, all three researchers watched the video clip together and identified the key features of the movement which led them to their decision (Davis et al., 2011). From this exercise, a final consensus score was achieved, and these raw scores were then averaged across repeated trials for each participant. Three other scores were determined based on these raw averaged scores: the worst averaged scores (highest score between legs), the best averaged scores (lowest score between legs) and binary averaged scores (pass – negative test: <2 and fail – positive test: ≥2 for each leg).

For the OLST trials, a measure of pelvic drop was determined using the custom kinematic model created for each participant in Visual3D. Graphical information reflecting the three-dimensional motion (flexion-extension (X), lateral bend (Y), and axial twist (Z) axes) of the pelvis was also used. While observing the kinematic model from the frontal and sagittal planes, the presence/absence of pelvic drop was identified based on a drop in the pelvis on the contralateral side (relative to the stance leg) which may or may not have been accompanied by trunk flexion towards the ipsilateral side, hip adduction, and internal rotation of the contralateral side (Kendall et al., 2010; Magee, 2007; Sahrmann, 2002). The presence/absence of pelvic drop was represented by pass/fail outcomes for each of the eight OLST trials (present = fail, coded as a '1'; absent = pass, coded as a'0'). For each participant, the pass/fail scores for each trial were then combined into one score three different ways: version 1 (overall average), version 2 (overall binary outcome), and version 3 (overall *adjusted* binary outcome). In version one (overall average), an average of all of the pass/fail scores was determined. For example, if pelvic drop was present in six of the eight trials, the overall average for this participant would be 0.75, with an outcome closer to 1 and 0 representing more and less presence of pelvic drop, respectively. For version two (overall binary outcome), a fail in one or more trials was considered to be an overall fail. If all trials were passed, this was considered to be an overall pass. In version three (overall *adjusted* binary outcome), a fail in two or more trials was considered to be an overall fail. This means that an overall pass in version three allows for a fail in one trial. Thus, participants had a greater chance of passing in version three. The OLST and SRT balance times were averaged across repeated trials for each participant and then relative change values were calculated (Equation 1).

Three VAS scores were recorded during standing. One VAS score was recorded immediately prior to standing (baseline: 0 minutes), and two VAS scores were recorded during standing (mid-point: 15 minutes and end-point: 30 minutes). To be able to link the LBP reported during standing to the standing exposure itself, the baseline VAS was subtracted from the mid- and end-points of standing (Nelson-Wong & Callaghan, 2010a; Schinkel-Ivy et al., 2013). By removing the bias (zeroing the VAS scores), all participants began the standing with a VAS score of 0 mm. In addition to the raw (bias removed) scores, maximum and mean VAS scores were also determined. A change of 10 mm from baseline was used as the cut-point to distinguish between which participants were classified as PD (>10 mm) and NPD (≤10 mm) (Gallagher et al., 2011; Nelson-Wong et al., 2009; Nelson-Wong & Callaghan, 2010a).

4.6 Data Processing for the Relationship between the Clinical Tests, and with Low Back Pain

To assess the relationship between the sidelying test (AHAbd test) and the standing balance tests (OLST, SRT), performance outcomes (scores for the AHAbd test and times for the standing balance tests) from visit one were used. A preliminary statistical analysis was initially run to investigate whether these data should be separated based on the factors of sex, side, and/or eye condition (eyes closed/eyes open) when applicable. Accordingly, associations were processed for the eyes closed and eyes open standing balance test (OLST, SRT) conditions separately. First, for each participant, binary pass/fail scores were determined for every trial for each test. Then, to be able to compare performance on the sidelying test with performance on the standing balance tests using Fisher's exact test (see Section 4.8), for each participant, the pass/fail scores for each trial were combined into one score two different ways. In version one (overall binary outcome), a fail in one or more trials was considered to be an overall fail (coded as a '1'). If all trials were passed, this was considered to be an overall pass (coded as a '0'). In version two (overall *adjusted* binary outcome), a fail in two or more trials was considered to be an overall fail. This means that an overall pass in version two allows for a fail in one trial. Thus, participants had a greater chance of passing in version two. For each of the AHAbd test trials, a pass was a score of

0 or 1 and a fail was a score of 2 or 3 on a four-point ordinal scale of 0-3. For each of the OLST trials and SRT trials, a pass/fail outcome was determined based on the achievement of maximum (30 seconds) and sub-maximum (<30 seconds) time, respectively.

To determine the association between the three clinical tests (AHAbd test and standing balance tests) and perceived LBP during standing, test performance outcomes (scores/times) and maximum VAS scores from visit one were used. Based on the preliminary statistical analyses run, associations were processed for the eyes closed and eyes open standing balance test (OLST, SRT) conditions separately. Test performance outcomes for the AHAbd test included the worst averaged scores (highest score between legs) and for the standing balance tests included balance times which were averaged across trials.

4.7 Data Analysis for the Insoles

All statistical analyses were performed using IBM SPSS Statistics v.20 (IBM Corporation, Armonk, USA) and an alpha level of 0.05 was used to indicate statistical significance. For analysis of variance (ANOVA), the most commonly used factors and their associated levels are as follows: visit (two visits), side (two levels: R/L, which reflects the lifted leg for the AHAbd test and stance leg for the OLST), and sex (two levels: male, female). Bonferroni corrections were used for post-hoc testing where significant interactions from ANOVAs were observed. When the assumption of sphericity was violated with Mauchly's test, the degrees of freedom were adjusted using Greenhouse-Geisser corrections. Further, for ANOVAs involving the factors of visit, sex, and/or side, when there were no significant main effects or interactions of sex, and/or side, data were collapsed across each of these factors.

4.7.1 Electromyography

To assess the effects of the insoles, EMG values were analyzed using ANOVA for each muscle separately for all three clinical tests. For the AHAbd test, maximum EMG values were analyzed in a three-way mixed-factor ANOVA with repeated measures of visit, side, and a between-group factor of sex. Maximum and mean relative change EMG values for the OLST were also analyzed using a three-way mixed-factor ANOVA (same statistical model as the AHAbd test), and for the SRT, were analyzed using a two-way mixed-factor ANOVA with a repeated measure of visit and a between-group factor of sex. Instrumentation failure (i.e. the electrodes lost contact with the skin) prevented the analysis of EMG data for the left EO (female: n=1) for the AHAbd test, and the left ES (female: n=1) and the right EO (male: n=1) for both the OLST and SRT.

4.7.2 Kinematics

To assess the effects of the insoles, three-dimensional kinematics were analyzed using ANOVA for each region (lumbar, pelvis, lumbopelvic, and trunk) separately for the OLST. Maximum and mean relative change angles for each axis (flexion-extension: X, lateral bend: Y, and axial twist: Z) were analyzed separately and combined (Euclidean norm angle) in a three-way mixed-factor ANOVA with repeated measures of visit and side, and a between-group factor of sex.

4.7.3 Kinetics

To assess the effects of the insoles, kinetic measures were analyzed using ANOVA for each COP measure (COP_d, COP_{RMS}, $COP_{totalpath}$, and COP_v) separately for the OLST and SRT. When applicable some COP measures (COP_d, COP_{RMS}, and COP_v) were analyzed in each axis separately. Relative change COP measures (absolute mean and peak-to-peak COP_d, COP_{RMS}, $COP_{totalpath}$, and absolute mean COP_v) were analyzed using a three-way mixed-factor ANOVA with repeated measures of visit and side, and a between-group factor of sex for the OLST, and a two-way mixed-factor ANOVA with a repeated measure of visit and a between-group factor of sex was used for the SRT. Instrumentation failure prevented the analysis of OLST and SRT kinetic data for one male participant.

4.7.4 Test Performance Outcomes and Perceived Low Back Pain

To assess the effects of the insoles, performance on the AHAbd test (rated scores), OLST (balance time and pelvic drop), and SRT (balance time) were analyzed using ANOVA. Some AHAbd test score versions (raw averaged scores, binary scores) and relative change OLST balance times were analyzed using a three-way mixed-factor ANOVA with repeated measures of visit and side, and a between-group factor of sex. Other AHAbd test score versions (best and worst scores), all OLST pelvic drop versions (average, overall binary, and overall *adjusted* binary outcomes), and relative change SRT balance times were analyzed using a two-way mixed-factor ANOVA with a repeated measure of visit, and a between-group factor of sex. Technical difficulties with the videotaped AHAbd test data prevented the analysis of AHAbd test scores for one male participant.

To determine the effects of the insoles on perceived LBP during standing, raw (bias removed) VAS scores were analyzed using a two-way ANOVA with repeated measures of visit and time point (baseline, mid-point and end-point of standing). Maximum and mean VAS scores were analyzed using a one-way ANOVA with a repeated measure of visit.

4.8 Data Analysis for the Relationship between the Clinical Tests, and with Low Back Pain

To assess the relationship between performance on the sidelying AHAbd test and the standing balance tests (OLST, SRT) during visit one, a Fisher’s exact test was run using 2 x 2 contingency tables to compare the proportion of participants who passed/failed the AHAbd test and who passed/failed the standing balance tests (OLST/SRT). This was done using two versions of the scores (overall binary outcome and overall *adjusted* binary outcome) for the eyes closed and eyes open standing balance test (OLST, SRT) conditions separately.

To quantify the relationship between performance on the three clinical tests (the AHAbd test, OLST, and SRT), and perceived LBP (VAS scores) during standing, correlations were performed. A Pearson product moment correlation was conducted for ratio data (OLST and SRT average balance times), and Spearman's rank-order correlation was conducted for ordinal data (worst AHAbd test scores between legs). Correlation coefficients were classified as very weak (0.00-0.19), weak (0.20-0.39), moderate (0.40-0.59), strong (0.60-0.79), and very strong (0.80-1.00) (Swinscow, 1997).

4.9 Data Analysis for the Active Hip Abduction Test Rater Reliability

Intraclass correlation coefficients (ICCs) were used to determine the inter-rater and intra-rater reliability of the AHAbd test scores in the present study. Five versions of the scores (raw, raw averaged, worst averaged, best averaged, and binary averaged scores – described previously) were used for these analyses. Inter-rater reliability was determined by comparing the scores between the three raters (C.S.B and two graduate students) using the scores from the first round of rating. These scores were analyzed using a two-way mixed-effects model with single measures consistency definition ($ICC_{3,1}$ model) (Shrout & Fleiss, 1979). The scores from the first and second rounds of rating were used to determine intra-rater reliability for each of the three raters individually using the same model ($ICC_{3,1}$). An average of the resulting intra-rater ICCs was calculated to provide an overall average intra-rater reliability score (Davis et al., 2011). To interpret the resulting ICCs, the following guidelines were used: 0.00-0.25 (little/if any correlation), 0.26-0.49 (low correlation), 0.50-0.69 (moderate correlation), 0.70-0.89 (high correlation), and finally, 0.90-1.00 (very high correlation) (Munro, 2005).

CHAPTER 5

Results

5. Results

Classification of PDs and NPDs will be discussed first, followed by descriptive information regarding participant activity levels and insole use. Next, results addressing research question one (the effects of the insoles) and two (the relationship between the three clinical tests, and these tests with perceived LBP) will be reported in order.

When reporting the results addressing the effects of the insoles, higher to lower order interactions are reported first (when applicable), followed by main effects. Relative change values were used to quantify the effects of the insoles in the majority of the OLST and SRT analyses to compare between the eyes closed and eyes open conditions (Equation 1). When interpreting the relative change results, both *absolute magnitude* and *sign* must be considered.

For all of the OLST and SRT relative change values, the interpretation of the *absolute magnitude* is the same: a larger number (further from zero) indicates a greater difference between the eyes closed and eyes open condition values, while a smaller number (closer to zero), indicates a lesser difference. Understood in the context of the insoles examined within the present study, a greater and lesser difference reflect worse and improved balance, respectively, suggesting changes in proprioception. When interpreting the *sign* for the OLST and SRT EMG, kinematic (OLST kinematics: Euclidean norm angle only), kinetic, and balance time relative change values, positive values indicate that the value for the eyes closed condition is greater than that of the eyes open condition, and the converse for values with a negative sign. However, for the OLST kinematics separated by axis, the *sign* reflects the direction of motion for each axis. Positive/negative values for the flexion-extension (X), lateral bend (Y), and axial twist (Z) axes reflect that the region of interest was more flexed/extended, laterally bent right/left, and axially twisted right/left, respectively for the eyes closed condition compared to the eyes open condition.

It is important to note the following when considering the interpretation of the *sign* for all of the standing balance test (OLST, SRT) relative change data. The relative change value accounts for the size (magnitude) of the eyes closed and eyes open values relative to the eyes open value. For example, a difference of 1 between an eyes closed value of 2 and an eyes open value of 1 would result in a relative change of 100%, while this same difference for an eyes closed value of 51 and an eyes open value of 50 would result in a relative change value of 2%. Thus, the relative of change of 100% is more functionally meaningful than the relative change of 2%. Thus, when a positive/negative sign reflects that they eyes closed condition was greater/lesser than the eyes open condition (for EMG, OLST kinematics: Euclidean norm angle only, kinetic, and balance times) or that the eyes closed condition was more flexed/extended, laterally bent right/left, and axially twisted right/left than the eyes open condition (for the OLST kinematics: separated by axis), this particularly reflects the functional meaning of the direction (sign) of the difference.

5.1 Pain Developer and Non-Pain Developer Classification

The transient LBP model has been used previously to differentiate between PDs and NPDs during prolonged standing. In the current study, a low percentage of participants were identified as PDs in visit one and two (20% – Table 5), compared to previous work which identified much higher percentages of PDs (40-71%: Gallagher et al., 2011; Marshall et al., 2011; Nelson-Wong & Callaghan, 2010abc; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008; Raftry & Marshall, 2012).

Table 5: The frequency (n and % of row total) of pain developers (PDs) and non-pain developers (NPDs) in visits one and two.

Visit	Sex	Pain Developer (PD)	Non-Pain Developer (NPD)
1	Female	2 (18.18%)	9 (81.82%)
	Male	2 (22.22%)	7 (77.78%)
	Total	4 (20.00%)	16 (80.00%)
2	Female	2 (18.18%)	9 (81.82%)
	Male	2 (22.22%)	7 (77.78%)
	Total	4 (20.00%)	16 (80.00%)

Given this low percentage, coupled with the relatively small sample size, the PD/NPD results were not used as a grouping factor for statistical analyses. Descriptively, there were a few interesting points of note. All of the PDs in visit one, and half of the PDs in visit two reached their PD status at the 30 minute mark of standing. This suggests that transient LBP development of a clinically relevant (PD status) level mostly seemed to start occurring at the end of the standing stimulus. In terms of the insoles effects on PD status, while the number (4) and breakdown (2 males and 2 females) of PDs was the same for visit one and two, only one male participant remained as a PD in visit two (see Table 5). The three other PDs (1 male and 2 females) from visit one became NPDs in visit two, and three NPDs (1 male and 2 females) from visit one became PDs in visit two. Taken together, the insoles appeared to produce mixed results, with no change, a worsening (change to PD status) and an improvement (change to NPD status) in the clinically relevant level of perceived LBP (PD status) reported by participants.

5.2 Effects of the Insoles

5.2.1 Participant Physical Activity Levels and Insole Use

Questionnaire data revealed that on average, participants spent the majority of their time sitting (mean ± SD: 6.26 ± 1.86 hours/day), followed by standing (2.90 ± 3.08 hours/day) walking (1.71

± 1.01 hours/day), and activities of daily living (1.04 ± 0.58 hours/day) during the week preceding visit one. Participants spent the least amount of their time engaging in resistance (0.89 ± 0.83 hours/day) and aerobic (0.80 ± 1.10 hours/day) exercise. When broken down by sex, females spent more time sitting (6.43 ± 1.95 vs. 6.02 ± 1.84 hours/day) and performing activities of daily living (1.08 ± 0.55 vs. 0.99 ± 0.70 hours/day) compared to males. Males spent more time than females standing (3.47 ± 3.98 vs. 2.45 ± 2.25 hours/day), walking (1.75 ± 1.21 vs. 1.67 ± 0.91 hours/day), and performing resistance (1.23 ± 0.88 vs. 0.58 ± 0.68 hours/day) and aerobic (1.11 ± 1.63 vs. 0.55 ± 0.31 hours/day) exercise. Taken together, participants appeared to spend most of their time in the week preceding visit one engaging in sedentary and light activity, especially females.

On average, participants wore the insoles for 6.49 ± 2.69 hours/day and at the end of the eight weeks of use, progressed to level 4.15 ± 1.76 of the six level foot-stimulating tab strengthening system. This may suggest that participants took longer (~ 2 weeks) to adjust to the insoles than anticipated based on the manufacturer’s guidelines (~ 4-7 day expected adjustment period). Males wore the insoles for less time (5.83 ± 3.02 vs. 7.08 ± 2.35 hours/day), and progressed to a lower level of the strengthening system than females (4.11 ± 1.90 vs. 4.18 ± 1.72). Taken together, male participants were less compliant with their use of the insoles than females based on their duration of use and level achieved.

5.2.2 Electromyography

Active Hip Abduction Test

Overall, no trunk, hip, or thigh muscles exhibited a significant change in maximum EMG values during the AHAbd test from visit one to visit two (p>0.066). The analysis of a mean value may have been helpful to detect changes across visits, as it reflects the EMG for the entire duration of the test, and therefore may be more sensitive, compared to just the peak portion of the test which is captured by the maximum. However, due to methodological limitations, only the maximum values were analyzed (see Section 7.). With no muscles reflecting a change in maximum EMG values across visits, the insoles did not appear to have an effect on the AHAbd test EMG.

One-Leg Stance Test

Overall, for the maximum relative change EMG results, only two thigh muscles, the left BF, and right VM exhibited a significant change during the OLST across visits. There was a significant three-way interaction between visit, side, and sex for the left BF muscle (F(1,18)=6.876, p=0.017, Figure 8). When males performed the left side test, the activation of the left BF muscle was higher during the eyes closed than the eyes open condition for both visits, with approximately a three-fold reduction (153.39%) in the relative change (improved proprioception) from visit one to visit two (226.14 ± 158.55% vs. 72.75 ± 68.80%).

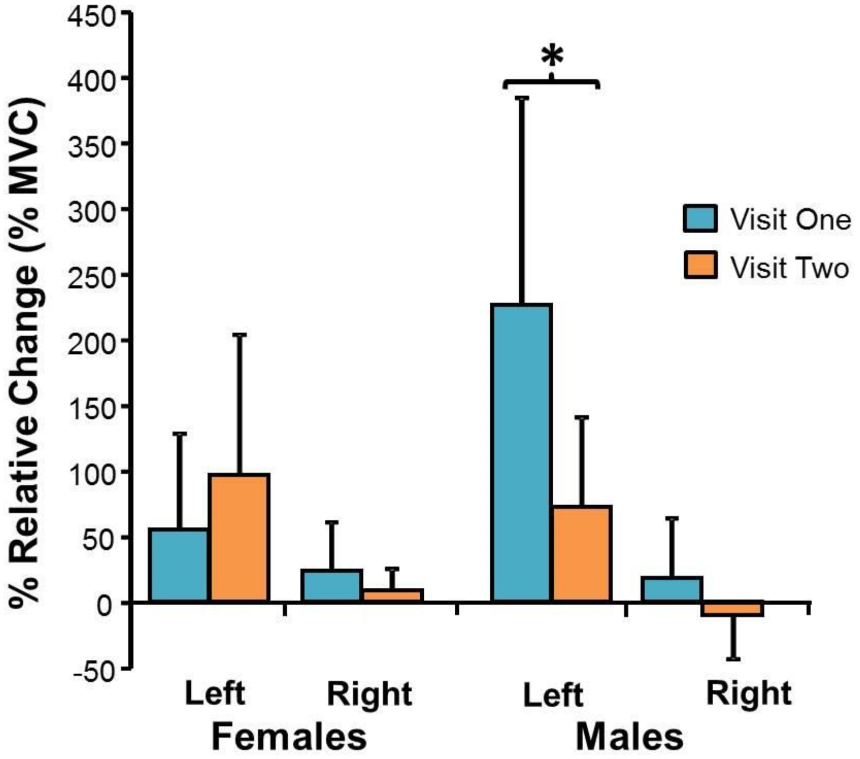

Figure 8: EMG activation of the left biceps femoris (BF) muscle during the one-leg stance test (OLST). There was a significant visit*side*sex interaction in which the decrease in relative change suggests an improvement in proprioception with insole use in visit two compared to visit one in males when the test was performed on the left leg. Values indicate the % relative change of the maximum EMG activation, as normalized to the %MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at $^*p<0.05$.

A significant main effect of visit was observed for the right VM muscle ($F(1,19)=6.002$, $p=0.024$, Figure 9). The activation of the right VM muscle was higher during the eyes closed than the eyes open condition for both visits, with over a 1.5 times reduction (44.50%) in the relative change (improved proprioception) from visit one to visit two (109.58 ± 100.54% vs. 65.08 ± 72.09%). Both muscles reflected an improvement in proprioception across visits. No significant main effects or interactions involving the factor of visit were observed for the remaining trunk, hip, and thigh muscles ($p>0.055$).

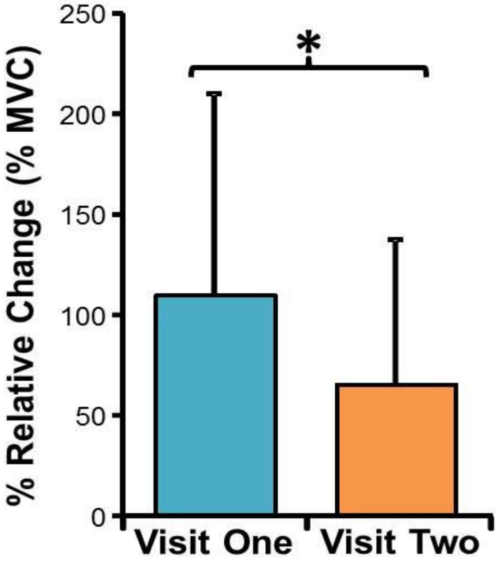

Figure 9: EMG activation of the right vastus medialis (VM) muscle during the one-leg stance test (OLST). There was a significant main effect of visit in which the decrease in relative change suggests improvements in proprioception with insole use in visit two compared to visit one. Values indicate the % relative change of the maximum EMG activation, as normalized to the %MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at $^*p<0.05$.

Overall, for mean relative change EMG results, only one trunk muscle, the left IO exhibited a significant change during the OLST from visit one to visit two, with a significant main effect of visit observed ($F(1,18)=9.933$, $p=0.006$, Figure 10). The activation of the left IO muscle was higher for the eyes closed than the eyes open condition for both visits, with approximately a three-fold increase (16.36%) in the relative change (worse proprioception) from visit one to visit two (7.80 ± 19.09% vs. 24.16 ± 28.41%). In contrast to the significant maximum relative change EMG results (left BF and right VM muscles), the left IO muscle exhibited a worse proprioception across visits. No significant main effects or interactions involving the factor of visit were identified for the remaining trunk, hip, and thigh muscles ($p>0.053$).

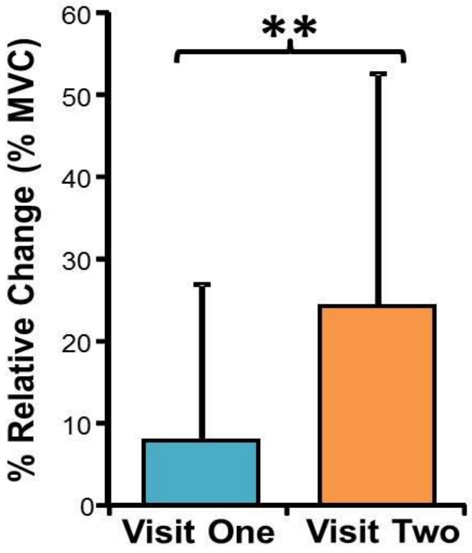

Figure 10: EMG activation of the left internal oblique (IO) muscle during the one-leg stance test (OLST). The left IO displayed a main effect of visit in which the increase in relative change suggests worse proprioception with insole use in visit two compared to visit one. Values indicate the % relative change of the mean EMG activation, as normalized to the %MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at ***p*<0.01.

Taken together, the insoles mostly had no effect on the maximum and mean relative change EMG for the OLST, with only two thigh muscles (left BF and right VM) reflecting improvements in proprioception.

Sharpened Romberg Test

Overall, for the maximum relative change EMG results, two trunk (left RA and left IO), one hip (right GTMED), and one thigh (left VM) muscle exhibited a significant change during the SRT from visit one to visit two. An interaction between visit and sex was observed for the left IO muscle (*F*(1, 18)=6.020, *p*=0.025, Figure 11). For males, the activation of the left IO muscle was higher during the eyes open than the eyes closed condition for visit one and the converse was true for visit two. There was approximately a 6.5 times increase (43.92%) in the relative change (worse proprioception) from visit one to visit two (-8.00 ± 24.20% vs. 51.92 ± 85.04%).

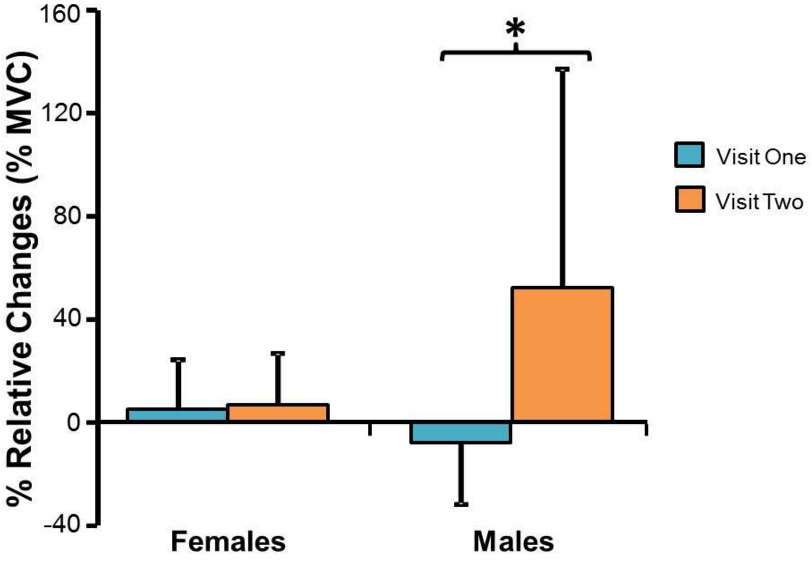

Figure 11: EMG activation of the left internal oblique (IO) muscle during the Sharpened Romberg test (SRT). There was a significant visit*sex interaction in which the increase in relative change suggests worse proprioception with insole use in visit two compared to visit one for males. Values indicate the % relative change of the maximum EMG activation, as normalized to the %MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at $*p<0.05$.

A main effect of visit was observed for the left RA, ($F(1,19)=5.171$, $p=0.035$), right GTMED ($F(1,19)=5.067$, $p=0.036$), and left VM ($F(1,19)=4.988$, $p=0.038$) muscles (Figure 12). For the right GTMED and left VM muscles, the activations were higher in the eyes closed than the eyes open conditions for both visits. Both muscles exhibited an increase in the relative change (worse proprioception) across visits (right GTMED: 31.09 ± 50.75% vs. 69.31 ± 70.83%; left VM: 39.40 ± 44.30% vs. 86.86 ± 105.67%), with over a two-fold increase (right GTMED: 38.22%; left VM: 47.46%). For the left RA muscle, the activation was higher for the eyes open than the eyes closed condition for visit one, and the converse was true for visit two. The left RA muscle also exhibited an increase in the relative change (worse proprioception) from visit one to visit two (-1.49 ± 26.75% vs. 7.60 ± 19.85%), with approximately a five-fold increase (6.11%).

Taken together, all four of these muscles (with all but one from the left side of the body) exhibited worse proprioception across visits. No significant main effects or interactions involving the factor of visit were observed for the remaining trunk, hip, and thigh muscles ($p>0.187$).

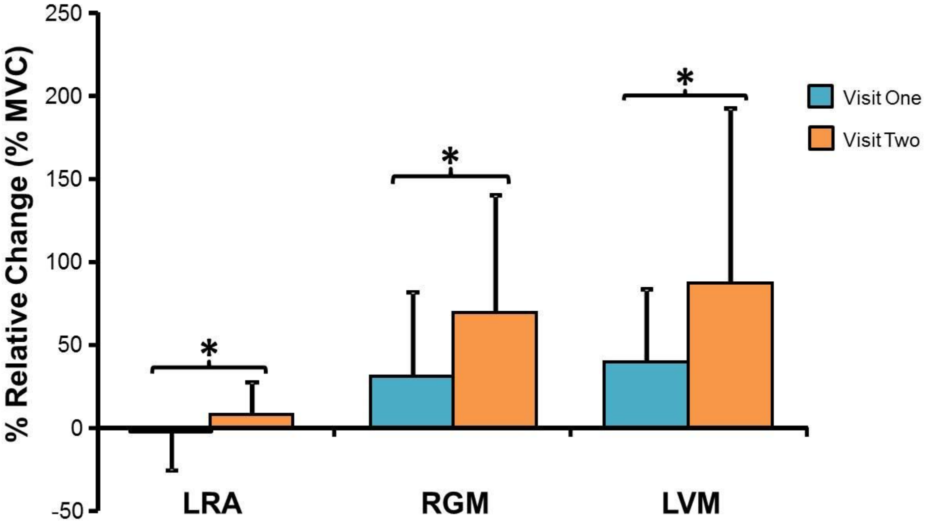

Figure 12: EMG activation of the left rectus abdominis (RA), right gluteus medius (GTMED), and the left vastus medialis (VM) during the Sharpened Romberg test (SRT). Each muscle showed a main effect of visit in which the increases in relative change in visit two suggest worse proprioception with insole use. Values indicate the % relative change of the maximum EMG activation, as normalized to the % MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at $^*p<0.05$.

Overall, for the mean relative change EMG results, three trunk (R/L LES and left RA) and one thigh (right BF) muscle exhibited a significant change during the SRT from visit one to visit two. A significant interaction between visit and sex was observed for the right ($F(1,18)=8.570$, $p=0.009$) and left ($F(1,17)=17.110$, $p=0.001$) LES (Figure 13) and right BF ($F(1,18)=4.556$, $p=0.047$) muscles (Figure 13). The activation of these three muscles was higher in the eyes closed than the eyes open condition for all of the significant pairwise comparisons. For the right LES muscle, males exhibited over a three-fold increase (58.42%) in the relative change (worse proprioception) from visit one to visit two (24.58 ± 27.12% vs. 83.00 ± 88.87%). Males and females exhibited opposite trends for the left LES muscle, with over a two-fold increase (24.41%) in the relative change (worse proprioception), and over a two-fold reduction (29.46%) in the relative change (improved proprioception), respectively, across visits (males: 18.70 ± 13.61% vs. 43.11 ± 31.52%; females: 52.18 ± 66.02% vs. 22.72 ± 49.03%).

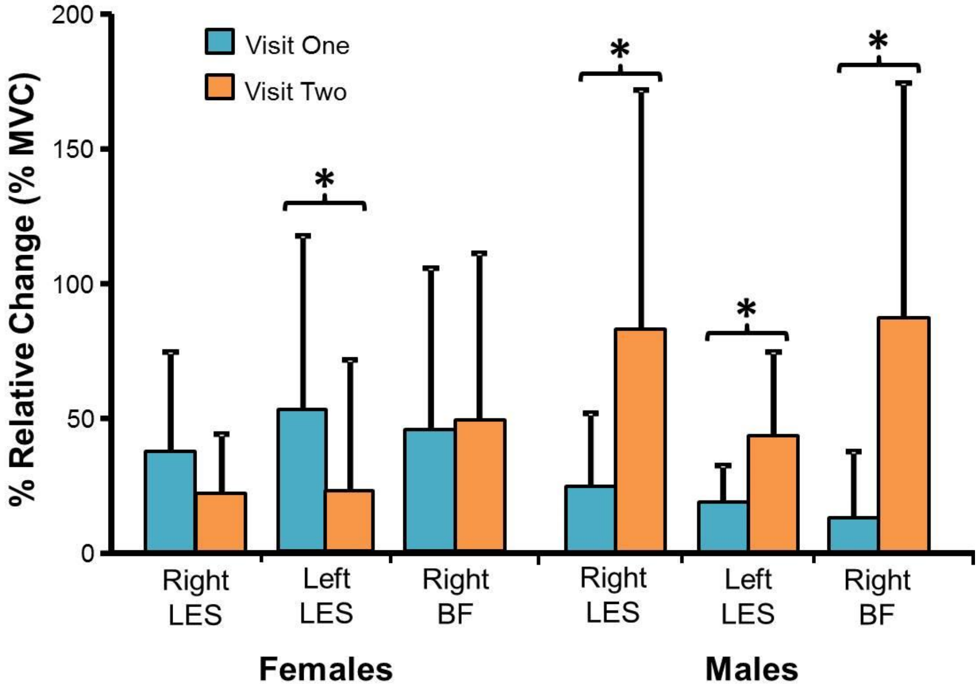

Figure 13: EMG activation of the R/L lumbar erector spinae (LES), and the right biceps femoris (BF) during the Sharpened Romberg test (SRT). All muscles showed a significant visit*sex interaction. In males, there was an increase in the relative change across muscles in visit two compared to visit one suggesting that insole use resulted in worse proprioception. In females however, the left LES showed a decrease in relative change in visit two compared to visit one which was associated with improved proprioception. Values indicate the % relative change of the maximum EMG activation, as normalized to the % MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at $*p<0.05$.

For the right BF muscle, males exhibited over a 6.5 times increase (74.24%) in the relative change (worse proprioception) from visit one to visit two (12.93 ± 24.79% vs. 87.17 ± 87.20%). Thus, males exhibited a worsening in proprioception for all three of these muscles (R/L

LES and right BF), while females exhibited an improvement in proprioception for the left LES muscle. Finally, a main effect of visit was observed for the left RA muscle (*F*(1,18)=4.513, *p*=0.048, Figure 14), with a higher activation in the eyes open than the eyes closed condition for visit one, and the opposite for visit two. There was approximately a four-fold increase (0.99%) in the relative change of the left RA muscle, reflecting a worsening of proprioception across visits (-0.30 ± 4.68% vs. 1.29 ± 4.48%). With the exception of the left LES muscle for females, all three of these muscles exhibited worse proprioception across visits for males.

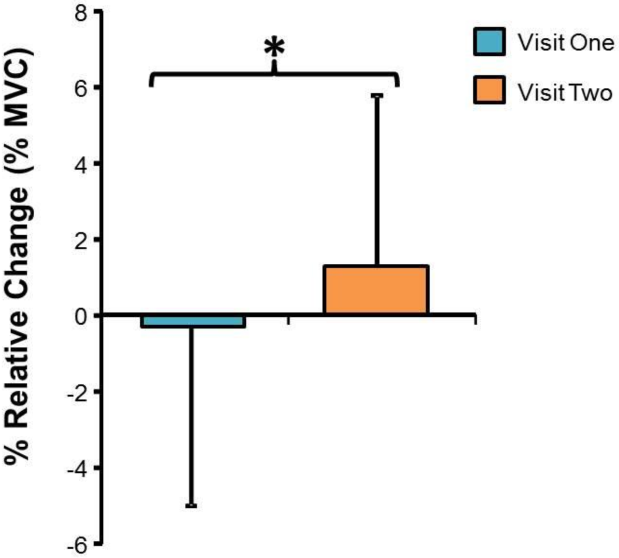

Figure 14: EMG activation of the left rectus abdominis (RA) during the Sharpened Romberg test (SRT). There was a main effect of visit in which the relative change increased in visit two compared to visit one suggesting that the use of the insoles worsened proprioception. Values indicate the % relative change of the mean EMG activation, as normalized to the % MVC, across the population with error bars reflecting standard deviation. Significant pairwise comparisons at **p*<0.05.

No significant main effects or interactions involving the factor of visit were identified for the remaining trunk, hip, and thigh muscles (*p*>0.088). Considered together, the insoles moderately impacted the maximum and mean relative change EMG results for the SRT, however, the majority of the effects reflected worse proprioception, particularly in males.

5.2.3 Kinematics

Overall, for the maximum and mean relative change angles in each axis separately, only the maximum lumbar angle in the lateral bend (Y) axis exhibited a significant change during the OLST from visit one to visit two, with an interaction between visit and sex observed ($F(1,18)=4.513$, $p=0.048$, Figure 15). For females, the lumbar posture exhibited during the visit one and two OLSTs reflected opposite patterns, with the eyes closed condition reflecting a more rightward and leftward lateral bend lumbar posture than the eyes open condition, respectively. There was approximately a 1.5 times (37.65%) increase in the relative change (worse proprioception) from visit one to visit two (63.37 ± 328.22% vs. -101.02 ± 242.27%). No significant main effects or interactions involving the factor of visit were identified for the remaining angles ($p>0.121$).

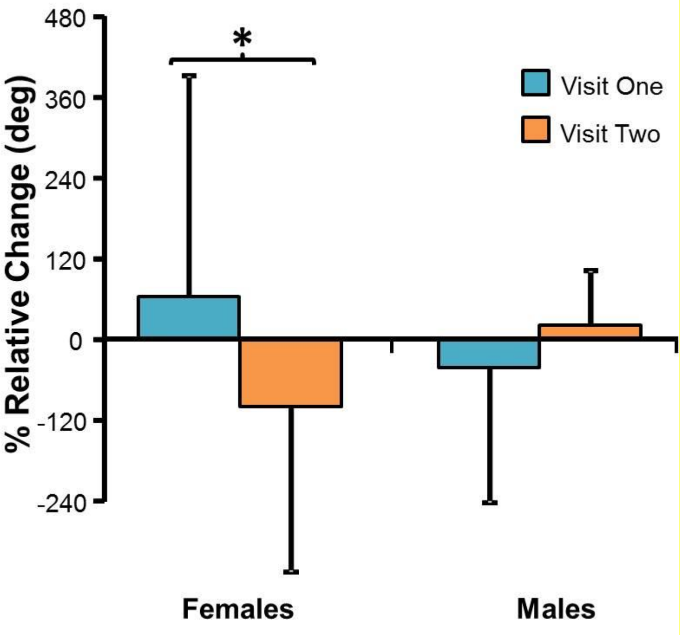

Figure 15: Kinematic angles for the lumbar region in the lateral bend axis during the one-leg stance test (OLST). There was a significant visit*sex interaction in which the relative change increased in the female population in visit two compared to visit one suggesting worse proprioception with insole use. Values indicate the % relative change of the maximum lumbar angle (deg) across the population with error bars reflecting standard deviation. Significant pairwise comparisons at $*p<0.05$.

Overall, for the maximum and mean Euclidean norm angle relative change results, only the maximum lumbopelvic Euclidean norm angle exhibited a significant change during the OLST across visits, with an interaction between visit and side observed (*F*(1,19)=5.234, *p*=0.034, Figure 16). For the left side test during visit one and two, the Euclidean norm angle for the lumbopelvic region was higher for the eyes closed than the eyes open condition. There was an eight-fold reduction (23.42%) in the relative change (improved proprioception) across visits (26.59 ± 47.38% vs. 3.17 ± 45.46%).

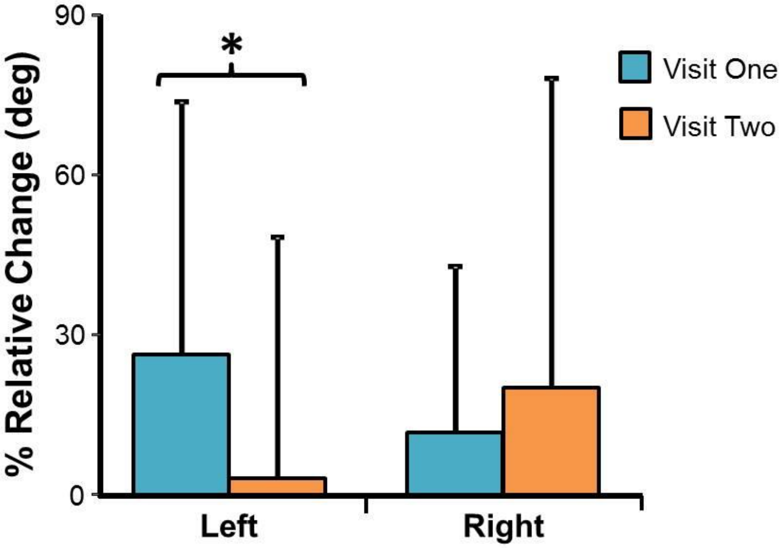

Figure 16: Euclidean norm angles for the lumbopelvic region during the one-leg stance test (OLST). There was a significant visit*side interaction in which the relative change decreased in the left leg in visit two compared to visit one suggesting worse proprioception with insole use. Values indicate the % relative change of the maximum Euclidean norm angle (deg) across the population with error bars indicating the standard deviation. Significant pairwise comparisons at $*p<0.05$.

No significant main effects or interactions involving the factor of visit were observed for the remaining maximum and mean Euclidean norm angles (*p*>0.353). Taken together, the insoles appeared to have very little impact on the kinematics of the four regions tested (trunk, pelvis, lumbar, lumbopelvic) with the only effects observed indicating equivocal (worse and improved proprioception) results.

5.2.4 Kinetics

One-Leg Stance Test

Overall, for the COP relative change results, only the peak-to-peak COP_d (M-L (Y) axis) and the COP_{RMS} (A-P (X) and M-L (Y) axes) exhibited a significant change during the OLST from visit one to visit two. A three-way interaction between visit, sex, and side was observed for the COP_{RMS} in the M-L (Y) axis ($F(1,17)=5.121$, $p=0.037$, Figure 17). For females during the right side test in both visits, the COP_{RMS} was greater for the eyes closed than the eyes open condition. There was over a 2.5 times reduction (72.20%) in the relative change (improved proprioception) from visit one to visit two (114.75 ± 4.84% vs. 42.55 ± 38.86%).

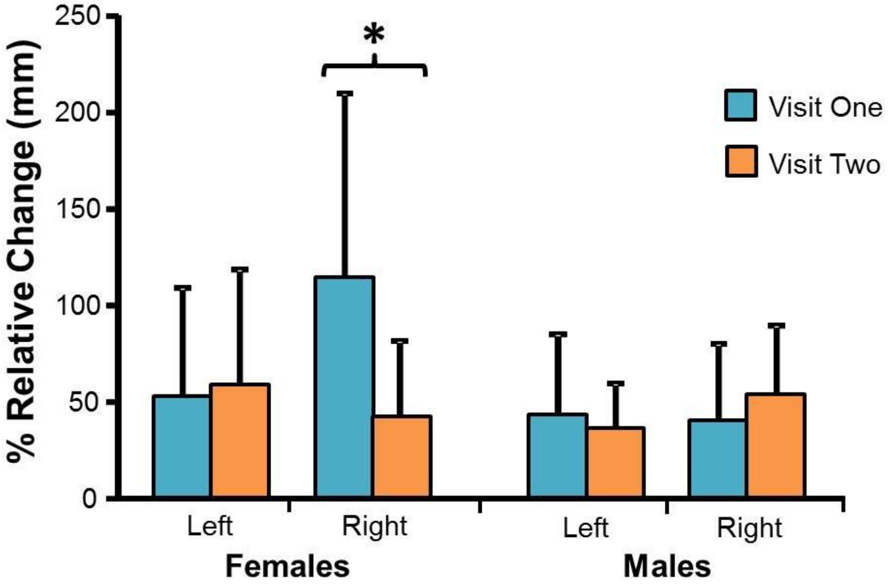

Figure 17: Centre of pressure root mean square (COP_{RMS}) for the medio-lateral (M-L) Y axis during the one-leg stance test (OLST). There was a significant visit*side*sex interaction in which the decrease in relative change suggests an improvement in proprioception with insole use in visit two compared to visit one in females when the test was performed on the right leg. Values indicate the % relative change of the COP_{RMS} (mm) across the population with error bars indicating standard deviation. Significant pairwise comparisons at $^*p<0.05$.

A significant main effect of visit was observed for the COP_{RMS} in the A-P (X) axis (*F*(1,18)=4.638, *p*=0.045, Figure 18A) and for the peak-to-peak COP_d in the M-L (Y) axis (*F*(1,18)=4.587, *p*=0.046, Figure 18B). These COP_{RMS} and the COP_d values were greater for the eyes closed than the eyes open condition for both visits, with a slight reduction (15.49%), in the relative change (improved proprioception) and over a five-fold reduction (15.05%) in the relative change (improved proprioception), respectively across visits (COP_{RMS}: 58.22 ± 26.18% vs. 42.73 ± 23.71%; COP_d: 18.56 ± 31.85% vs. 3.51 ± 12.80%). No significant main effects or interactions involving the factor of visit were observed for the remaining COP measures (p>0.083).

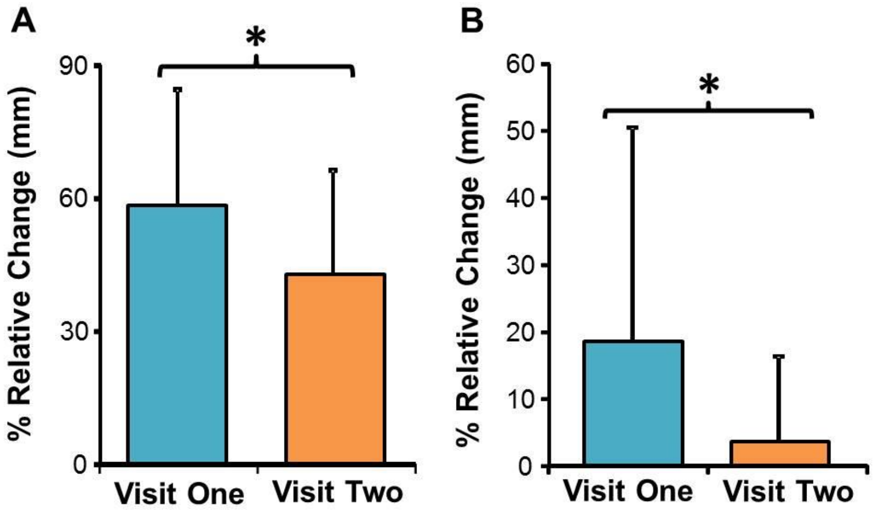

Figure 18: A) Centre of pressure root mean square (COP_{RMS}) for the antero-posterior (A-P) X axis and **B)** Centre of pressure displacement (COP_d) for the medio-lateral (M-L) Y axis, during the Sharpened Romberg test (SRT). Both showed a main effect of visit in which reduced relative change is reflective of worse proprioceptive performance with insole use. Values indicate the % relative change of the COP_{RMS} (mm) and COP_d (A and B respectively) with error bars reflecting the standard deviation. Significant pairwise comparisons at *p<0.05.

Taken together, the insoles only affected a few of the OLST COP measures, with all affected measures indicating improvements in proprioception.

Sharpened Romberg Test

No COP measures for the SRT exhibited a significant change from visit one to visit two (*p*>0.094). Thus, the insoles appeared to have no effect on any of the SRT COP measures.

5.2.5 Test Performance Outcomes and Perceived Low Back Pain

Overall, none of the test performance outcomes (AHAbd test scores, OLST and SRT relative change balance times, and OLST pelvic drop scores) exhibited a change from visit one to visit two (*p*>0.083). For the SRT relative change balance times, a significant interaction between visit and sex was observed, however, none of the pairwise comparisons were significantly different. Taken together, the insoles appeared to have no effect on any of the test performance outcomes for all three clinical tests.

Overall, no VAS scores exhibited a significant change from visit one to visit two (*p*>0.620). Not surprisingly, a significant main effect of time point was observed for the raw (bias removed) VAS scores was observed (*F*(1.327,25.212)=7.479, *p*=0.007). While there was over a four times higher (4.34 mm) VAS score at the end-point of standing compared to baseline, this is not considered clinically relevant (Hagg et al., 2003; Kelly, 1998). Taken together, the insoles appeared to have no effect on transient LBP development during standing.

5.3 Relationship between the Three Clinical Tests and with Perceived Low Back Pain

Overall, when examining the relationship between the sidelying AHAbd test and standing balance tests (OLST, SRT) in visit one, no significant associations were observed (*p*=1.000,Tables 6, 7, 8: frequencies and contingencies). No significant correlations were observed for any of the three clinical tests with maximum VAS scores during standing in visit one (*p*>0.085, Table 9). Taken together, it appears that none of the standing balance tests are related to the AHAbd test, and none of these three clinical tests are related to transient LBP development during standing.

Table 6: Frequencies presented as n (% of column total for each test) of the overall binary and overall *adjusted* binary pass/fail data for the active hip abduction (AHAbd) test, one-leg stance test (OLST), and Sharpened Romberg test (SRT) during visit one.

Test: Pass/Fail	**Overall Binary**	**Overall *Adjusted* Binary**
AHAbd test		
Pass	6 (30.00%)	11 (55.00%)
Fail	14 (70.00%)	9 (45.00%)
OLST Eyes Open		
Pass	17 (85.00%)	19 (95.00%)
Fail	3 (15.00%)	1 (5.00%)
OLST Eyes Closed		
Pass	1 (5.00%)	6 (30.00%)
Fail	19 (95.00%)	14 (70.00%)
SRT Eyes Open		
Pass	20 (100.00%)	20 (100.00%)
Fail	0 (0.00%)	0 (0.00%)
SRT Eyes Closed		
Pass	10 (50.00%)	20 (100.00%)
Fail	10 (50.00%)	0 (0.00%)

Table 7: Contingency table for the overall binary and overall *adjusted* binary pass/fail data for the active hip abduction (AHAbd) test and one-leg stance test (OLST) during visit one. Data is presented as n (% of row total), *p*=1.000.

Version	**AHAbd test**	**OLST Eyes Open** Pass	**OLST Eyes Open** Fail	**Total**
Overall Binary	Pass	5 (83.30%)	1 (16.70%)	6
	Fail	12 (85.70%)	2 (14.30%)	14
	Total	17	3	20
Overall *Adjusted* Binary	Pass	10 (90.90%)	1 (9.10%)	11
	Fail	9 (100.00%)	0 (0.00%)	9
	Total	19	1	20
Version	**AHAbd test**	**OLST Eyes Closed** Pass	**OLST Eyes Closed** Fail	**Total**
Overall Binary	Pass	0 (0.00%)	6 (100.00%)	6
	Fail	1 (7.10%)	13 (92.90%)	14
	Total	1	19	20
Overall *Adjusted* Binary	Pass	3 (27.30%)	8 (72.70%)	11
	Fail	3 (33.30%)	6 (66.70%)	9
	Total	6	14	20

Table 8: Contingency table for the overall binary and overall *adjusted* binary pass/fail data for the active hip abduction (AHAbd) test and the Sharpened Romberg test (SRT) during visit one. Data is presented as n (% of row total), *p*=1.000.

Version	**AHAbd test**	**SRT Eyes Open** Pass	**SRT Eyes Open** Fail	**Total**
Overall Binary	Pass	6 (100.00%)	0 (0.00%)	6
	Fail	14 (100.00%)	0 (0.00%)	14
	Total	20	0	20
Overall *Adjusted* Binary	Pass	11 (100.00%)	0 (0.00%)	11
	Fail	9 (100.00%)	0 (0.00%)	9
	Total	20	0	20
Version	**AHAbd test**	**SRT Eyes Closed** Pass	**SRT Eyes Closed** Fail	**Total**
Overall Binary	Pass	3 (50.00%)	3 (50.00%)	6
	Fail	7 (50.00%)	7 (50.00%)	14
	Total	10	10	20
Overall *Adjusted* Binary	Pass	11 (100.00%)	0 (0.00%)	11
	Fail	9 (100.00%)	0 (0.00%)	9
	Total	20	0	20

Table 9: Correlation coefficients between the three clinical tests, the active hip abduction (AHAbd) test worst averaged scores, one-leg stance test (OLST) average balance times, Sharpened Romberg test (SRT) average balance times and maximum perceived LBP visual analogue scale (VAS) scores. Pearson product moment correlation unless indicated - Spearman's rank-order correlation[†].

Test and Associated Measure	**Correlation Coefficient**	**Classification**	***p*-value**
AHAbd test Worst Averaged Scores	0.04	Very weak	0.861[†]
OLST Eyes Open Average Balance Times	-0.40	Moderate	0.085
OLST Eyes Closed Average Balance Times	-0.28	Weak	0.233
SRT Eyes Open Average Balance Times	-0.05	Very weak	0.822
SRT Eyes Closed Average Balance Times	0.16	Very weak	0.499

5.4 Active Hip Abduction Test Rater Reliability

Inter-rater and intra-rater reliability were determined for the raw, raw averaged, worst averaged, best averaged, and binary averaged AHAbd test scores using ICCs (SEM: standard error of measurement) (Table 10). Overall, for all of these versions, the inter-rater reliability was mostly moderate, ranging from moderate, 0.57 (0.49), to high, 0.75 (0.51), and the intra-rater reliability for individual raters was mostly high, ranging from low, 0.44 (0.43), to high, 0.77(0.36) as shown in Table 11 (Munro, 2005). When the ICCs for individual raters were averaged across the raters for each version, the outcomes reflected a similar range to the individual scores, ranging from low, 0.49 (0.43) to high, 0.75(0.42). Considering all of the versions of the AHAbd test scores that were analyzed, the highest and lowest inter-rater reliability was observed for the worst averaged and binary averaged scores, respectively and the highest and lowest intra-rater reliability was observed for the best averaged and binary averaged scores, respectively. Thus, for both inter – and intra – rater reliability, the binary averaged scores were the least reliable. Taken together, the inter- and intra- rater reliability were identified to be mostly moderate and high for the AHAbd test scores, indicating that within the current study, this test was found to be reliable.

Table 10: Inter-rater reliability of each of the three raters for the raw, raw averaged, worst averaged, best averaged, and binary averaged active hip abduction (AHAbd) test scores using intraclass correlation coefficients (ICCs) and standard error of measurement (SEM).

Version	$ICC_{3,1}$	Classification	SEM
Raw	0.64	Moderate	0.65
Raw Averaged	0.67	Moderate	0.52
Worst Averaged	0.75	High	0.51
Best Averaged	0.60	Moderate	0.47
Binary Averaged	0.57	Moderate	0.49

Table 11: Intra-rater reliability of the three raters for the raw, raw averaged, worst averaged, best averaged, and binary averaged active hip abduction (AHAbd) test scores using intraclass correlation coefficients (ICCs) and standard error of measurement (SEM).

Version	Rater	$ICC_{3,1}$	Classification	SEM
Raw	1	0.67	Moderate	0.48
	2	0.60	Moderate	0.58
	3	0.68	Moderate	0.54
	Mean	0.65	Moderate	0.53
Raw Averaged	1	0.68	Moderate	0.40
	2	0.70	High	0.41
	3	0.72	High	0.42
	Mean	0.70	High	0.41
Worst Averaged	1	0.73	High	0.45
	2	0.76	High	0.38
	3	0.76	High	0.42
	Mean	0.75	High	0.42
Best Averaged	1	0.58	Moderate	0.34
	2	0.75	High	0.31
	3	0.77	High	0.36
	Mean	0.70	High	0.34
Binary Averaged	1	0.44	Low	0.43
	2	0.58	Moderate	0.37
	3	0.44	Low	0.48
	Mean	0.49	Low	0.43

CHAPTER 6

Discussion

6. Discussion

This thesis had two major aims centred on lumbopelvic control. The first involved an investigation of the effects of insoles on lumbopelvic control using biomechanical (muscle activation, three-dimensional kinematic, kinetic), and test performance outcomes (rated scores/times/pelvic drop) during three clinical tests including a sidelying test of lumbopelvic control (the AHAbd test), and two standing balance tests (the OLST and SRT). The second was to examine the relationship between the sidelying test, and two standing balance tests. As a part of these two major aims, it was also of interest to assess the effects of the insoles on perceived LBP, and to determine the association between the three clinical tests and perceived LBP. With a lack of studies in the insoles literature focusing on the lumbopelvic region, and as the first study to investigate the relationship between a lumbopelvic control test and standing balance tests, this thesis aimed to address these gaps within the literature, with potential implications for targeted LBP prevention and intervention strategies.

While insoles have been widely prescribed by clinicians to treat LBP (Walter et al., 2004), the mechanisms by which these devices work are still not fully understood. Using a comprehensive set of biomechanical measures and clinically relevant tests, this thesis aimed to help clarify which lumbopelvic control measures (i.e. EMG, kinematics, kinetics) and which clinical test (the AHAbd test, OLST, SRT) may be the most responsive to the insoles investigated. The use of a relative change equation to quantify biomechanical outcomes for the balance tests was unique in that this type of analysis is lacking within the insoles literature, yet powerful in its interpretative value. Specifically, this type of analysis allows researchers to draw conclusions about insoles with an emphasis on somatosensory (proprioceptive) mechanisms (that may improve lumbopelvic control) which are of particular interest for these devices. Further, by determining whether there is an association between a sidelying test of lumbopelvic control, which is moderately predictive of LBP during standing, and two standing balance tests, which both heavily rely on weight-bearing lumbopelvic control, this thesis aimed to identify whether these balance tests are related, yet more functionally relevant clinical markers of LBP, compared to the sidelying test.

Overall, the insoles did not appear to be effective in improving any of the biomechanical and test performance outcomes examined for all three clinical tests, and no change in perceived LBP was observed. There also appeared to be no association between the sidelying test of lumbopelvic control and two standing balance tests, as well as no association between all of these tests and perceived LBP.

6.1 Pain Developer and Non-Pain Developer Classification

The results of the transient LBP development model must first be addressed to explain why participants were not grouped based on the PD/NPD classification for statistical analysis. While there is a wide range of PDs reported in previous work using this model (40-71%: Gallagher et al., 2011; Marshall et al., 2011; Nelson-Wong & Callaghan, 2010abc; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008; Raftry & Marshall, 2012), the percentage of PDs in the present study was well below this, at 20%. As the sample size for the present study was relatively small (n = 20), only four of twenty participants make up this 20%. This is markedly lower than the 17 of 43 participants who make up the lowest percentage (40%) of PDs observed in previous work (Nelson-Wong & Callaghan, 2010abc; Nelson-Wong et al., 2009). This was not sufficient, especially when considering that the majority of statistical analyses were 2 and 3-way mixed model ANOVAs. Taken together, a higher number of PDs was required to have included the

PD/NPD classification as a factor within the statistical analyses. Limitations which likely prevented a higher number of PDs within the present study are discussed in Section 7.

6.2 Effects of the Insoles on Biomechanical and Test Performance Outcomes

6.2.1 Electromyographical Measures

Overall, the insoles were not effective in improving the trunk, hip, and thigh EMG activity for all three major clinical tests, with mixed effects observed for both balance tests (the OLST and SRT), and no effects observed for the AHAbd test. This is evident when considering that in total, the insoles only significantly altered 12 of 80 possible different EMG output measurements (64 of which were from the balance tests: 16 muscles x 2 output metrics x 2 balance tests, and 16 of which were from the AHAbd test: 16 muscles x 1 test x 1 output metric). Although most of the significant changes were observed during the SRT (9/32 EMG outputs), they were almost exclusively reflective of worse proprioception (increased relative change). Conversely, the few significant changes observed during the OLST (3/32 EMG outputs) were mostly reflective of improved proprioception (decreased relative change). The variability of these findings is in line with previous studies which also observed equivocal EMG outcomes when examining some of the same trunk (Bird et al., 2003), hip (Bird et al., 2003; Dingenen et al., 2015; Hertel et al., 2005), and thigh (Dingenen et al., 2015; Hertel et al., 2005; Mündermann et al., 2006; Nawoczenski & Ludewig, 1999) muscles during various tasks (balance, walking, running, and selected exercises) following insole and heel lifts/foot wedges (Bird et al., 2003) use. That the present study found changes at all in the EMG activity of the muscles it did (trunk: R/L ES; hip: right GTMED; and thigh: R/L BF and R/L VM), is also in line with this previous body of work.

What remains unclear after reviewing this literature however, is what constitutes an improvement or worsening of EMG activity (increase/decrease amplitudes or earlier/later onsets). In addition, there is no standard for what is considered an improvement or worsening in EMG outcomes with insole use, and therefore, it is difficult to compare across studies (Murley et al., 2009). This is further complicated by the fact that EMG is highly variable within and between individuals, and across different tasks (De Luca, 1997). However, when a comparable single-limb balance test was examined (Dingenen et al., 2015), despite methodological differences (i.e. participants with chronic ankle instability; months of custom insole use due to musculoskeletal problems, and the use of EMG timing as a parameter etc.), the significant change in the VM muscle, and not the GTMED muscle during the transition to single-limb stance in the current study, is in agreement with the findings of Dingenen and colleagues (2015).

To the author's knowledge, the present study is the first to analyze the effects of insoles on EMG relative change values during balance tests, thus, no direct comparisons with the literature can be made. However, despite the dearth of studies which investigate the effects of insoles on balance outcomes using EMG activity, previous research assessing balance using kinetic COP outcomes have suggested that insoles enhance the somatosensory (tactile and proprioceptive) information on the plantar surface of the foot (Hamlyn et al., 2012; Hijmans et al., 2007). This enhanced somatosensory input has been postulated to contribute to improvements in neuromuscular control of the foot and whole body which is critical for maintaining balance and posture during standing (Hamlyn et al., 2012; Hijmans et al., 2007; Mochizuki & Amadio, 2003; Olmstead & Hertel, 2004; Rothbart, 2005). The insoles in the present study are designed to target somatosensory mechanisms (using a dome-shaped arch and progressively larger/firmer foot-stimulating tabs), thereby improving neuromuscular and postural control, and in turn, reducing LBP. By removing sensory information with an eyes closed balance condition, coupled with the use of tests which require a small (OLST) and narrow

(OLST, SRT) BOS, the postural context of the present study demanded a heavy reliance on, and challenge to, the somatosensory system which was of particular interest. With this sensory re-weighting which emphasizes somatosensory inputs (Carver et al., 2006), and the insoles theoretically enhancing proprioception in the eyes closed condition, this would be expected to reduce the difference (relative change) between the eyes closed and open conditions, with the EMG activity in the eyes closed condition (condition of worse postural control) being more reflective of the eyes open condition (condition of better postural control) (Horak & Macpherson, 1996; Romberg, 1851). This comparison of eyes closed and eyes open conditions was achieved through the use of a relative change equation in the present study, where the role of proprioception can be more easily discriminated from other sensory inputs (Morioka et al., 2000). This is in contrast with previous researchers who were more limited in their ability to make this discrimination, without using this comparison, and instead using EMG amplitude and timing outputs alone. While the present study, and previous studies which investigated the effects of insoles on balance did not directly quantify proprioception, the strength of the inferences made using whole body postural control lie in the manipulation of the postural context (and/or analyses) to emphasize somatosensory inputs.

When examining the specific muscles which significantly changed with insole use in the present study, while most of the improvements (SRT: ~29%; OLST: ~45-154%) and worsening (SRT: ~24-74% excluding the left RA; OLST: ~16%) in proprioception appeared to be functionally relevant in terms of magnitude, no strong and consistent patterns in the muscles identified were apparent within and across the balance tests. Despite both improvements during the OLST occurring in thigh muscles (left BF, right VM), this change was only observed for the maximum values (which reflects a peak moment in the test), and not for the mean values (which reflects muscle activation through the duration of the test) in the same muscles. Additionally, the large improvement (~153% reduction in the relative change) in the left thigh muscle (BF) did not seem to translate proximally to the trunk (left IO muscle), which actually reflected a worsening with insole use. This is somewhat surprising considering the structural and functional connection of the BF muscle to the lumbopelvic region (via its proximal attachment at the pelvis (Drake et al., 2012; Tortora & Nielsen, 2012) and its contribution to spinal stability and whole body postural control (McGill, 2007; Neumann, 2010a). The large improvement observed in this left thigh muscle (for males) may be explained by the fact that it occurred during the left leg test when proprioceptive mechanisms would be more pronounced (left leg was weight-bearing) compared to the right leg test (left leg was non-weight-bearing).

The significant changes during the SRT were identified in mostly maximum values for muscles from all three regions (trunk, hip, and thigh) with only the left RA muscle exhibiting changes in both maximum and mean values. Not only were most of the changes reflective of worse proprioception with insole use, but this worsening was always observed in males when sex contributed to these results. This was particularly interesting for the left LES muscle which produced conflicting findings, with males and females exhibiting worse and improved proprioception, respectively. It is possible that all of the worse results in males may be attributed to their lower compliance with their insole use (less hours/day wearing them and lower level of progression in the strengthening system) relative to females.

When considering the significant changes for both balance tests (OLST, SRT), the left IO was the only muscle which was identified to change in both tests, in the same direction (both worse proprioception as reflected by an increased relative change). Further, all of the changes in the abdominal muscles across both tests reflected this same pattern of worsening. This may indicate that this proximal region may not have neuromuscularly adjusted to the insoles after eight weeks of use. Perhaps earlier or later measurements of participants may have revealed very different EMG outcomes taking into account the time it takes for neural adaptations to translate in the motor control system (Behm, 1995). Slightly more improvements were observed in the OLST compared to the SRT, possibly due to the greater somatesensory demands (smaller and narrower BOS) required during the OLST (Palmieri et al., 2002). Specifically, these somatosensory demands may make this test more responsive to enhanced proprioception, especially in a young, unaffected population, with insole use. Taken together, these results suggest the insoles were ineffective and produced highly variable results for the trunk, hip, and thigh EMG activity during the balance tests examined.

For the AHAbd test, none of the 16 muscles examined exhibited a change in magnitude from visit one to visit two. To the author's knowledge, no studies have examined the muscle activation amplitudes for the AHAbd test in general, or as a result of insole use. However, the results observed in the present study are in line with magnitudes observed for similar tasks including sidelying hip abduction (Bolgla & Uhl, 2005), and the ASLR test (Liebenson, Karpowicz, Brown, Howarth, & McGill, 2009). A possible explanation for the lack of significant changes with this test may be due to the fact that it is performed with the eyes open and in a sidelying position. As the insoles selected for the present study are specifically thought to cause improvements through proprioceptive mechanisms, tests which are performed with the eyes closed (no visual input), and therefore rely more heavily on somatosensory inputs (Carver et al., 2006; Horak & Macpherson, 1996), may be more sensitive to possible changes in proprioception. Without the plantar surface of the feet in contact with the ground as a result of the non-weight-bearing sidelying position required for this test, the proposed somatosensory changes resulting from the insoles may have been less pronounced compared to weight-bearing standing positions. Taken together, for all three major clinical tests examined, especially the AHAbd test, the insoles were not effective in improving EMG activity outcomes.

6.2.2 Kinematic Measures

Compared to the EMG outcomes, the insoles had even less impact on the kinematics of the trunk, pelvis, lumbopelvic, and lumbar regions examined during the OLST, with the few changes observed reflecting mixed results. Only two of 32 possible different kinematic output measurements (24 of which were from the separated axes: 4 regions x 3 axes x 2 output metrics, and 8 of which were from the combined axes Euclidean norm angles: 4 regions x 2 metrics) were significantly altered as a result of insoles use. Specifically, only the lumbar (for the separated axes) and lumbopelvic (for the Euclidean norm angles) regions exhibited changes in maximum values across visits. While both significant changes were functionally relevant, the lumbar region reflected worse proprioception (increased relative change of ~38%), while the lumbopelvic region reflected improved proprioception (reduced relative change of ~23%) during the OLST. Very few studies have examined the effects on insoles on the kinematics of regions proximal to the shank, and those that did, analyzed these regions during gait (Esfandiari et al., 2013; Marinakis & Catalfamo, 2004; Nester et al., 2003). While methodological differences make it difficult to draw comparisons, the minimal and lack of effects of insoles on the trunk (Esfandiari et al., 2013; Marinakis & Catalfamo 2004) and pelvis (Marinakis & Catalfamo, 2004; Nester et al., 2003) regions observed in this literature are generally in agreement with the present study. As no studies to the author’s knowledge have investigated the effects of insoles on kinematics of these proximal regions during balance tests, and the present study is the first, to the author’s knowledge, to do this using relative change analyses, no direct comparisons can be made with the literature. Other research however, has reported kinematic data for upright standing, and in comparison to this work, kinematic values in the eyes open and eyes closed conditions alone, fall within the reported ranges of this literature (Gallagher, Wong, & Callaghan, 2013; Pearcy, Portek, & Shepherd, 1984; Pearcy & Tibrewal, 1984; White & Panjabi, 1978).

When examined separately by axis, only the lumbar region changed across visits demonstrating a reduction in proprioception (increased relative change) exclusively for females during the OLST. For both visits, the lumbar spine appeared to be in a more laterally deviated posture (more rightward in visit one and more leftward in visit two), with a greater deviation observed in visit two. Based on the proposed improvements in proprioception that may have resulted from the use of the insoles, to have been considered beneficial, the lumbar spine should have assumed a more upright (less deviated) posture. The more laterally deviated posture reflects decreased postural control, and compared to a more upright posture, may produce an increased risk of injury in terms of LBP (O'Sullivan et al., 2003).

When overall motion was assessed (Euclidean norm angle), only the lumbopelvic region reflected a change across visits during the OLST. It exhibited an improvement in proprioception (decreased relative change) when the participants performed the OLST on their left leg, which for 90% of participants, was their non-dominant leg. As the Euclidean norm angle provides an estimate of combined three-dimensional motion, this suggests that the lumbopelvic region exhibited less motion in general during visit two compared to visit one. This reduction in lumbopelvic motion is thought to reflect an improvement in postural control (Horak & Macpherson, 1996) for this young unaffected population. Based on the theoretical kinematic bases which have been formulated within the insoles literature, and the enhancement of proprioceptive mechanisms as a result of insole use, it is possible that this decreased lumbopelvic motion may have resulted from kinematic changes in the foot and lower extremities, which propagated up the kinematic chain (Bird & Payne, 1999). However, this appears unlikely to have occurred in the current study, as the improvement in the lumbopelvic region (T_{12}-pelvis) was not replicated in the lumbar spine (T_{12}-L_5), which is contained within this region, and thus would be expected to produce similar results. Further, no significant changes were observed in the pelvis segment alone. When taken together, the minimal kinematic results which are conflicting (improvements/worsening) for regions that are anatomically and functionally linked, strongly suggest that the insoles were ineffective in altering and improving the kinematics of the regions examined.

6.2.3 Kinetic Measures

Overall, while the insoles had little impact on all of the biomechanical measures examined, the most improvements occurred in kinetic COP outcomes during the OLST. No improvements were observed during the SRT. Three of 18 possible different kinetic output measurements (1 COP parameter x 2 axes x 2 output metrics + 2 COP parameters x 2 axes x 1 output metric + 1 COP parameter x 1 combined axis x 1 output metric x 2 balance tests) were significantly changed during the OLST as a result of insole use. All of these changes during the OLST were reflective of improved proprioception (reduced relative change) resulting from insole use. Only one of these changes was observed in the A-P (X) axis (COP_{RMS}), while the remaining two changes were observed in the M-L (Y) axis (peak-to-peak COP_d, COP_{RMS}). This suggests that the insoles seemed to be most effective in enhancing frontal plane postural control. In terms of magnitude, the most functionally relevant reduction in relative change was observed for the COP_{RMS} in the M-L (Y) axis (~72%), while the other two improvements (COP_{RMS} in the A-P (X) axis and peak-to-peak COP_d in the M-L (Y) axis) were substantially less functional (~15%).

Despite the mixed results within the insoles literature, the results of the present study generally fall in line with previous work which has also observed reductions in COP measures (improved balance) during bilateral (Losa Iglesias et al., 2012; Qiu et al., 2012; Qui et al., 2013) and single-leg stance (Hamlyn et al., 2012; Olmstead & Hertel, 2004) balance tests following insole use. However, while the specific COP measures to show improvements during single-leg stance tests were not all examined within the present study (i.e. COP area), the significant improvement in COP_v in previous work (Olmstead & Hertel, 2004) is not in line with the present study which observed no change in this specific COP measure. While no researchers, to the author's knowledge, have examined the effects of insoles on COP outcomes during the SRT, as improvements have been observed for more (OLST) and less (bilateral stance) challenging balance tests, some improvements in COP outcomes during the SRT test were expected.

In line with the EMG results for the present study, it is possible that changes were observed for the OLST and not the SRT because of the greater somatosensory demand (smaller and narrower BOS) of the OLST stance which may make it a more sensitive test with regard to proprioceptive changes, especially for a young unaffected population which may already be performing near optimal levels of postural control (Hamlyn et al., 2012; Palmieri et al., 2002). Additionally, as the present study did not control for different foot types, which have been shown to have an effect on balance COP outcomes (Hertel, Gay, & Denegar, 2002), the ability of the present study to detect additional improvements may have been confounded by this factor. Taken together, the insoles used in the present study were minimally effective in improving kinetic outcomes for the OLST and not effective for the SRT.

6.2.4 Test Performance Outcomes and Perceived Low Back Pain

Overall, the insoles had no impact on any of the performance outcomes examined (AHAbd test scores, OLST and SRT relative chance balance times, OLST pelvic drop scores) for all three clinical tests and had no impact on perceived LBP during standing. There are several potential explanations for this. Based on the lack of significant motor control (EMG activity) changes during the AHAbd test in the present study, it is not surprising that there were no significant changes in observed lumbopelvic movement control (rated scores) as a result of insole use. The lack of a weight-bearing posture (no plantar proprioceptive input) for this test may mean that if any changes did occur with insole use, they may have been much less pronounced than those of weight-bearing postures (with the presence of plantar proprioceptive input). When compared to the only work (Nelson-Wong, 2009) which investigated the effects of an intervention on the AHAbd test, while there were several methodological differences (i.e. the use of a trunk stabilization exercise intervention, the lack of repeated AHAbd test trials bilaterally), the absence of significant changes in scores across visits in the present study falls in line with this previous work.

The lack of changes observed with the OLST and SRT balance times in the present study was somewhat surprising. This was based on the proposed somateosensory mechanism and balance improvements purported to be associated with the insoles examined, and considering the targeted manipulation of the postural context (eyes closed, narrow/small BOS) and relative change analysis to emphasize the somatosensory system. All considered, participants were expected to exhibit improved (higher) balance times during the eyes closed condition which relies more heavily on somatosensory inputs (vs. eyes open; Horak & Macpherson, 1996), ultimately resulting in a lower relative change value (greater similarity between the eyes closed and eyes open conditions) (Losa Iglesias et al., 2012). Unlike with some biomechanical measures (i.e. EMG), the expected values at baseline, and the interpretation (improvements/worsening) of balance times is very clear in a young, unaffected population for eyes closed and eyes open conditions: balance times are shorter in an eyes closed condition, and an increased balance time constitutes improved performance, while a decreased balance time constitutes worse performance (Springer et al., 2007). Thus, improvements as a result of insoles would have been abundantly clear with balance times, yet were not observed, reflecting the ineffectiveness of these devices for this measure.

Research investigating pelvic drop is generally sparse, and to the author's knowledge, is not performed within the insoles and balance literature. The lack of significant changes in this OLST measure for the present study may be attributed to a few factors. While pelvic drop is used as a common clinical measure during the Trendelenberg test, and has been postulated to have associations with LBP, the very little literature which has examined this measure is equivocal in its findings (Penney et al., 2014). The prevalence of pelvic drop and, thus its potential for improvements in the young unaffected population examined within this study are unknown, as there appears to be no solid basis for comparison within this literature. Further, two methodological considerations must be discussed. As binary pass/fail scores (absence/presence of pelvic drop) for the repeated OLST trials were combined into an overall outcome for each participant, this may have masked the effects that could have been observed for individual trials. Additionally, in a clinical setting, pelvic drop is typically identified by a clinician based on palpation and observation. However, in the present study, it was identified based on the visual inspection of custom three-dimensional kinematic models for each participant. As a result, it is uncertain whether the presence/absence of pelvic drop was correctly identified based on the lack of palpation and/or kinematic quantification.

Based on the mostly minimal perceived LBP development during standing in the present study, it is not surprising that there were no significant changes in the maximum and mean VAS scores with insole use. To have improved this measure (a decrease in VAS scores), a modest to high level of perceived LBP during standing in visit one likely needed to be present, to allow for potential changes to occur. Taken together, the insoles were not effective in improving the test performance outcomes and perceived LBP during standing within the present study.

6.3 Relationship between the Three Clinical Tests and with Perceived Low Back Pain

Overall, the sidelying AHAbd test exhibited no association with any of the standing balance tests (OLST, SRT) examined, and none of these tests exhibited associations with transient LBP development during standing in visit one. Methodologically, the examination of the association between the sidelying and standing balance tests may have been limited in a few ways. The use of combined scored data from repeated trials may have washed out potential effects that may have been observed when just comparing individual trials which were not transformed. While this did not seem to be apparent based on the insoles EMG results, it is possible that these tests may have exhibited a relationship if biomechanical measures were compared instead. While no previous work has attempted to associate the sidelying AHAbd test to more functionally relevant, upright standing balance tests, the lack of association between these tests appears to generally fall in line with the work of Olivier and colleagues (2015). When investigating the association between lumbopelvic control and single-leg standing balance tests with injury incidence in a population of cricket athletes, while the lumbopelvic control tests were not performed in a sidelying position, and the lumbopelvic control and balance tests were not directly associated, these tests did not appear to be related within this study which only identified single-leg standing to be associated with injury. In line with the lack of significant changes in transient LBP during standing within the insoles analyses, the absence of a relationship between the scored data for these tests and maximum VAS scores is most likely attributed to the very minimal levels of transient LBP that were induced in the present study. Taken together, it appears that none of the standing balance tests are related to the AHAbd test, and none of these three clinical tests are related to transient LBP development during standing.

6.4 Active Hip Abduction Test Rater Reliability

In terms of the AHAbd test score reliability analyses, the ICCs observed in the present study appear to be reasonable when compared to previous researchers' findings (Davis et al., 2011). Using the same guidelines for interpretation of the ICCs (Munro, 2005), for inter-rater reliability performed on the raw (4-point ordinal scale) AHAbd test scores, the ICC for the present study was slightly lower than that observed by Davis et al. (2011), with moderate (0.64) compared to high (0.70) ICCs, respectively. When examining the intra-rater reliability ICCs performed on the raw (4-point ordinal scale) AHAbd test scores, the averaged ICCs for all raters was lower for the present study relative to Davis et al. (2011), with moderate (0.65) compared to high (0.74) ICCs, respectively. For these same scores (raw 4-point ordinal scale), the ICCs for individual raters was more consistent for the present study with all raters achieving a moderate ICC (0.60-0.68). However, for Davis et al. (2011), the ICCs for individual raters ranged from moderate to high (0.53-0.93).

Davis et al. (2011) did not complete reliability analyses on the four other versions of the AHAbd test scores which were examined within the present study (raw averaged, worst averaged, best averaged, and binary averaged), and thus, no direct comparisons can be made for these. Overall, the inter-rater reliability for these versions was mostly moderate, and the intra-rater reliability was mostly high. However, with mostly low intra-rater reliability, the binary averaged scores were the exception. It is likely that the ICCs observed in the present study were lower than those of Davis et al. (2011) as the raters in this previous work were practicing physical therapists who have more experience identifying subtle aberrant movement control differences that the three raters graduate student raters in the present study who do not have this same clinical experience. Further, Davis et al. (2011) obtained their ICC values from 16 raters, while the present study only included three. With a larger number of raters who are clinically trained to identify impairments in motion, the higher ICC values obtained in Davis et al. (2011) were expected.

CHAPTER 7

General Limitations

7. General Limitations

It is important to recognize the potential limitations associated with this study. Some of these were related to the instrumentation required to record kinematic, muscle activation, and kinetic signals. To minimize skin movement artefact (the deformation and displacement of the skin relative to the bone) (Leardini, Chiari, Della Croce, & Cappozzo, 2005), double-sided surgical tape was used to adhere the markers to the surface of the skin. While the signals recorded using surface EMG can be contaminated by improper electrode placement and crosstalk (the unintentional recording of activity from muscles nearby the muscle of interest) (De Luca 1997), previously defined, widely used electrode placement locations were selected for the present study. Although the surface EMG instrumentation (i.e. EMG leads, packs for batteries and leads, and straps to hold the packs in position) was quite encumbering for participants, adjustable straps were used to tailor the positioning and tightness of this instrumentation to each participant, and EMG leads were adhered to the skin to minimize cable sway artefact. These measures were taken to allow participants to function as comfortably and naturally as possible, while still maintaining the appropriate conditions to record the EMG signals. When performing a tandem stance on a force platform, participants' feet may exceed the length of the platform (Palmieri et al., 2002). This was also the case for the present study, such that participants had to position their feet diagonally (from corner-to-corner) to perform the SRT. While this means that the A-P (X) and M-L (Y) axes COP outcomes are not in line with conventional orthopedic axes, all participants were positioned with the same orientation on the force platform during both visits. Without time-synced kinematic data to accompany the surface EMG recordings during the AHAbd test, it was not possible to verify the start and end points of motion, which are not appropriate to obtain from EMG recordings alone (i.e. due to the electromechanical delay between the neural onset of electrical activity in the muscle and measureable tension in the EMG signal; Cavanagh & Komi, 1979). Thus, to be conservative, mean values were omitted from the AHAbd test EMG analyses as they very likely included data which was not representative of motion during the task.

There were a few limitations in terms of study design. While the sample size (n=20) used in the present study is relative small, it is comparable to other insoles studies (Dankerl et al., 2014; Ferrari, 2007), and is reasonable based on the challenges associated with recruitment of participants for a two visit, time intensive intervention study. Although the present study may have been strengthened by the inclusion of a control group, other insoles studies have also lacked a control group for comparison (Dananberg & Guiliano, 1999; Ferrari 2007), but have still advanced the understanding of insole mechanisms and effectiveness within the literature. By not accounting for the different foot types (i.e. high arch/flat footed) of participants, the current study did not control for a variable which may have affected the balance test COP outcomes (Hertel et al., 2002). As this study used a young, unaffected population, the results may not be generalizable to other populations (i.e. clinical populations). However, it is important to examine young unaffected populations such that the results may serve as a basis for comparison and may identify key considerations for when similar measures are used within clinical populations.

The low percentage of PDs in the present study was likely due to the much shorter standing stimulus duration of 30 minutes compared to two hours which was used by all of the previous work (Gallagher et al., 2011; Marshall et al., 2011; Nelson-Wong & Callaghan, 2010abc; Nelson-Wong et al., 2009; Nelson-Wong et al., 2008; Raftry & Marshall, 2012). This was further limited by the relatively small sample size used for the present study. Only 30 minutes was used as prolonged standing was not a major focus for the present study, which already included a comprehensive and time-intensive protocol. In the present study, most (75%)

PDs in visit one and two only developed their PD status at the end of the 30 minutes of standing. In previous work, PD status mostly seemed to onset in the second hour of standing, past 60 (Marshall et al., 2011) and 75 minutes (Nelson-Wong & Callaghan 2010ab; Nelson-Wong et al., 2009; Nelson-Wong et al., 2010). Taken in the context of previous work, this suggests that to achieve a PD/NPD split of 40%+, a standing stimulus greater than 60 minutes is likely required.

CHAPTER 8

Hypotheses Revisited

8. Hypotheses Revisited

This study examined the effects of insoles on lumbopelvic control during three clinical tests, a sidelying test of lumbopelvic control (the AHAbd test), and two standing balance tests (the OLST and SRT). The effects of this insole on perceived LBP during standing was also examined. Using trunk, hip, and thigh muscle activations (AHAbd test, OLST, and SRT), three-dimensional lumbopelvic kinematics (in each axis separately, and combined as a Euclidean norm angle – OLST), kinetic COP (displacement, root mean square, total path, and velocity – OLST and SRT), test performance outcomes (scores and/or times – AHAbd test, OLST, and SRT), and VAS scores (perceived LBP during standing), overall, the insoles did not appear to be effective in improving these measures during all three clinical tests. This study also investigated the relationship between the sidelying AHAbd test and upright OLST and SRT, as well as the relationship between these three tests and perceived LBP during standing. No associations were observed between the tests and with perceived LBP.

Hypothesis #1A states: Following eight weeks of use, insoles will decrease the maximum surface EMG activity of the trunk, hip, and thigh muscles during the AHAbd test, and will improve balance (likely due to proprioceptive mechanisms) during the OLST and SRT as reflected by a decrease in the difference (relative change) between the eyes closed and eyes open conditions for the maximum and mean surface EMG activity of these same muscles.

Decision: **Hypothesis rejected.**

With no changes in the activations of any of the muscles for the AHAbd test, the insoles did not alter the motor control of participants during this test of lumbopelvic control. During the OLST, two muscles did show improved proprioception with insole use as reflected in a decrease in the relative change in visit two. At first, this may suggest that the OLST is a sensitive measure until one considers that the insoles improved only two out of a possible 16 muscles. A number of muscles showed altered EMG activity in the SRT. However, in all but one instance, the activity suggested worse proprioception with insole use. Therefore the above hypothesis was rejected as stated above.

Hypothesis #1B states: Following eight weeks of use, insoles will improve balance (likely due to proprioceptive mechanisms) as reflected by a decrease in the difference (relative change) between the eyes open and eyes closed conditions for maximum and mean three-dimensional lumbar, pelvis, lumbopelvic and trunk angles (in each axis separately, and combined as a Euclidean norm angle) during the OLST.

Decision: **Hypothesis rejected.**

While two of the angles tested were altered due to insole use, in one case the alteration was reflective of improved proprioception and in the other case, was reflective of worse proprioception. Additionally, there were a possible 24 kinematic outputs that could have been altered and yet 22 of them remained unaffected by insole use. These minimal and mixed results suggest that overall, trunk and lumbopelvic kinematic measures were not sensitive to insole use. Thus this hypothesis was rejected as stated above.

Hypothesis #1C states: Following eight weeks of use, insoles will improve balance (likely due to proprioceptive mechanisms) as reflected by a decrease in the difference (relative change) between the eyes open and eyes closed conditions for COP (displacement, root mean square, velocity, and total path) during the OLST and SRT.

Decision: **Hypothesis rejected.**

While the insoles did not generally affect the COP measures examined, three outcomes did exhibit differences across visits during the OLST. No changes were observed for the SRT. Unlike with EMG and kinematic measures, there was a consistent improvement in proprioception observed due to insole use. However, these few positive COP outcomes should be considered within the context of the number of total outcomes tested (18 in total). Thus this hypothesis was rejected as stated above.

Hypothesis #1D states: Following eight weeks of use, insoles will improve performance on the AHAbd test as reflected through a decrease in the score, and will improve performance on the OLST and SRT (likely due to proprioceptive mechanisms) as reflected through a decrease in the difference (relative change) between the eyes open and eyes closed conditions for the balance times and a decrease in the presence of pelvic drop during the OLST.

Decision: **Hypothesis rejected.**

The insoles did not impact any of the performance outcomes for any of the three clinical tests. Thus this hypothesis was rejected as stated above.

Hypothesis #1E states: Following eight weeks of use, insoles will decrease perceived LBP (VAS scores) during standing.

<u>Decision:</u> Hypothesis rejected.

The insoles did not reduce perceived LBP (VAS scores) across visits. Thus this hypothesis was rejected as stated above.

Hypothesis #2 states: Performance on the AHAbd test (pass/fail based on scores) will be related to performance on the OLST and SRT (pass/fail based on balance times) during visit one, and performance on all three of these clinical tests (scores/balance times) will be correlated to maximum perceived LBP (VAS scores) during standing in visit one.

<u>Decision:</u> Hypothesis rejected.

No association was observed between performance on the AHAbd test and performance on the standing balance tests (OLST, SRT), and no association between the three clinical tests (AHAbd test, OLST, and SRT) and the maximum perceived LBP (VAS scores) during standing was observed. Thus, the hypothesis was rejected as stated above.

Impact

While the use of insoles as a means of improving proprioception, and in turn affecting lumbopelvic control, could have a large clinical impact on LBP sufferers, the minimal and often equivocal findings in the current study suggest that these particular insoles do not promote these types of neuromuscular changes. Of the number of possible outcome measures tested, very few showed alterations, and even fewer showed improved proprioception with insole use. In addition, the lack of a consistent pattern across and within muscles, kinematics, kinetics, or the tests examined, would suggest that the improvements in proprioception that *were* found are not indicative of these insoles being a useful intervention in lumbopelvic control mechanisms. Although standing balance tests would perhaps be a more functionally relevant assessment measure within populations that develop LBP due to prolonged standing, the current study suggests that the standing balance tests examined (OLST, SRT) do not present alternative assessment measures to the sidelying hip abduction test. However, it would be useful to test this assumption by using the extended transient LBP model that found associations between the sidelying hip abduction test and LBP.

CHAPTER 9

General Conclusions

9. General Conclusions

Most of the insoles literature heavily focuses on the lower extremities, with largely postulated links to the lumbopelvic region. With such a lack of empirical evidence for the biomechanical changes in the lumbopelvic region, this thesis aimed to address this gap by examining a comprehensive set of EMG, kinematic, and kinetic measures. In line with the insoles literature, the results of the present study appear to be quite equivocal with improvements, worsening, and no change in these measures being observed as a result of insole use. In addition, outcome measures that showed improved proprioception and lumbopelvic control were proportionally very few when compared to the number of outcomes tested. Due to the lack of literature surrounding the lumbopelvic region coupled with the use of the relative change equation for the balance test outcomes, it was difficult to make direct comparisons with previous research which is already heterogeneous. However, the use of relative change as a novel approach within this literature allows for more direct testing of the neuromuscular mechanisms that are thought to underlie improved lumbopelvic control with insole use. Specifically, relative change serves as a means of analytically reweighting somatosensory input more heavily towards proprioception, and thus, changes in output measures can more directly speak to whether lumbopelvic control is being affected through proprioceptive mechanisms. The current study provides a preliminary step towards filling a large gap within the insole literature and to forming a basis for comparison for future work which may investigate the lumbopelvic region. As there is strong evidence to suggest that impairments in the lumbopelvic region are associated with LBP with a bidirectional link to foot function, this region warrants further investigation in the insoles literature with clinical implications in terms of LBP prevention strategies. With no association observed between a sidelying test of lumbopelvic control (the AHAbd test) and upright standing balance tests (OLST, SRT), the application of this sidelying test to more functional, widely used clinical tests is not supported and may require further investigation using the extended transient LBP model.

CHAPTER 10

References

10. References

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Appendix

Appendix A: Segment Coordinate Systems

To define the four rigid segments in the kinematic model (trunk, upper and lower lumbar spine, and the pelvis), both anatomical and tracking markers were used (C-Motion Research Biomechanics Wiki-Documentation. Tutorial: Building a Model, http://c-motion.com/v3dwiki/ index.php? title=Tutorial:_Building_a_Model). Anatomical markers refer to markers that were placed in anatomically relevant locations including palpable bony landmarks and at each end of a segment. Tracking markers refer to the markers that were placed at convenient locations to track the segment. The anatomical markers on the right and left side of each segment end were arbitrarily defined as the lateral and medial markers, respectively. The segment endpoints were then calculated as the mid-point between the medial and lateral anatomical markers at the proximal and distal ends of the segment. Segment coordinate systems were calculated as described below (Constructing the Segment Coordinate System, http://c-motion.com/v3dwiki /index.php? title= Constructing_the_ Segment_ Coordinate _System; Schinkel-Ivy, 2015). The *z*-axis (supero-inferior) was defined as the vector passing through the proximal and distal segment endpoints, projecting upwards in the supero-inferior axis, in line with gravity. The frontal (*x*-*z*) plane was then calculated by fitting a leastsquares plane to the two proximal and two distal segment endpoints. The Y (A-P) axis was defined as a vector positioned orthogonally with respect to the frontal plane and Z axis, projecting forwards in the antero-posterior axis. The X (M-L) axis was then defined orthogonally with respect to the sagittal (Y-Z) plane and *y*-axis, projecting to the right of the participant in the X (M-L) axis.