EFFECTS OF AN EIGHT-WEEK INSOLE TRIAL PERIOD ON THE KINEMATICS AND MUSCLE ACTIVITY DURING THE STANCE PHASE WALKING

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Abstract

Insoles are currently used to alleviate and prevent low back pain, with some being designed to affect the kinematic and/or kinetic chain. The purpose of this study was to examine and quantify whether a neuromuscular training insole can alter muscle activation and kinematics during the stance phase of walking following two months of use. Eight males and eight females were given a neuromuscular training insole and attended collection sessions prior to the insole insertion and following eight weeks of insole use. While the insoles did not appear to have a large effect on muscle activation or lower limb and lumbar spine kinematics, they did appear to have an effect on thoracic spine movement. A reduction in thoracic spine mean, maximum, and minimum flexion angle was observed following the trial period; however, there was no indication this was a result of a change in lower limb kinematics. These findings indicated that these insoles appear safe for use and may serve to reduce thoracic spine flexion during walking.

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CHAPTER 1

Introduction

1. Global Introduction

Insoles have been prescribed to treat low back pain (LBP), despite the little evidence supporting the postulated mechanisms behind their success. Most of the research supporting the use of insoles to treat low back pain is solely questionnaire based (Dananberg and Guiliano, 1999; Ferrari, 2007; Cambrom et al., 2011), with little kinematic and EMG evidence quantifying their ability to alter biomechanics. There are currently multiple theories suggesting the mechanism behind their success in treating and preventing LBP, that range from shock absorption, to correcting for limb length discrepancy and the kinematic chain.

The theory of the kinematic chain is commonly investigated to determine the effects of insoles on the low back. The kinematic chain suggests that pronation of the foot during the stance phase of walking causes an increase in the internal rotation of the tibia and femur, finally resulting in anterior tilting of the pelvis and the lumbosacral joint (Bird and Payne, 1999). Many insoles are designed in attempt to reduce pronation at the foot, aiming to reduce the anterior tilting of the pelvis (Lafortune et al., 1994). While this has somewhat been displayed in standing research (Betsch et al., 2011), no changes were experienced by the spine and there is little evidence supporting this theory during walking. Studies have displayed a reduction in foot pronation and tibia internal rotation after insole use, with no major difference being found in knee, hip and pelvis kinematics (Marinakis and Catalfamo, 2004; Nester et al., 2003). It remains unclear whether insoles can alter spine kinematics via the kinematic chain.

Surface electromyography (EMG) has also been investigated in an attempt to uncover part of the mystery behind the success of insoles in treating LBP. Research by Tomaro and Burdett (1993), Ogon et al. (2001), Bird et al. (2003), and Murley and Bird (2006), displayed different changes in muscle activity, one increasing and one decreasing in activation.

Considering that changes have been documented in both directions it is important to establish the purpose of the insole before describing the change in EMG as beneficial or harmful. Nigg (2001) described how insoles encouraging an optimal movement path should act to reduce muscle activity. However, Nigg et al., (2006) describes an increase in muscle activity as a result of training/strengthening footwear as beneficial to improving strength, as well as proprioception. The intended function of the footwear, and whether it was designed for everyday or training purposes, must be taken into account before describing a potential change in muscle activation as beneficial or harmful.

1.1 Research Questions

The purpose of this study was to determine if an eight week neuromuscular training insole trial period could alter the three dimensional (3D) joint kinematics of walking, with a corresponding change in muscle activity of the lower limbs and torso. A secondary purpose was to determine if the insoles could change either joint angle or muscle activity, independent of the other. The following questions were addressed in this data collection

1. Could an eight week insole trial period change the observed range of motion (ROM), as well as mean, maximum, or minimum angle, for lower limb and spine kinematics, in a way that would support the theory of the kinematic chain reducing spine flexion?
2. Could an eight week insole trial period change the mean surface EMG activation observed for muscles of the lower limbs and torso?
3. Would changes experienced as a result of an eight week trial period manifest themselves in both muscle activation and joint angle data, or could it result in a change in only one of the two variables?

1.2 Hypotheses

Muscle activity and joint kinematics, for the lower limbs and torso, where observed before and after an eight week insole trial period. The following hypotheses were prepared in response to the above research question.

1. A decrease in internal rotation of the knee and hip, as well as a decrease in pelvic anterior tilt and spine flexion, will be detected following eight weeks of insole use (post-trial).
2. An increase in mean surface EMG activity will be detected as a result of the neuromuscular training insole.
3. The insoles will result in a decrease in mean joint angle that is related to a change in mean muscle activity of muscles responsible for moving said joint.

CHAPTER 2

Review of the Literature

2. Review of the Literature

The following is a review of the literature relevant to the lumbar and thoracic spine's motion during the stance phase of walking and some of the previous research examining insoles. In addition to the anatomy and movement patterns of the spine during walking, this review also examined the relevant anatomy and typical movement of the lower limbs, in order to explain the theory behind the kinematic chain. While certain aspects of the spine anatomy will be explained in depth, lower limb movement and anatomy will only cover the aspects that relate to the kinematic chain during stance phase. In addition to this, the typical methodology regarding the capturing of kinematics and muscle activation during walking are also examined.

2.1 Anatomy Literature Review

2.1.1 Foot

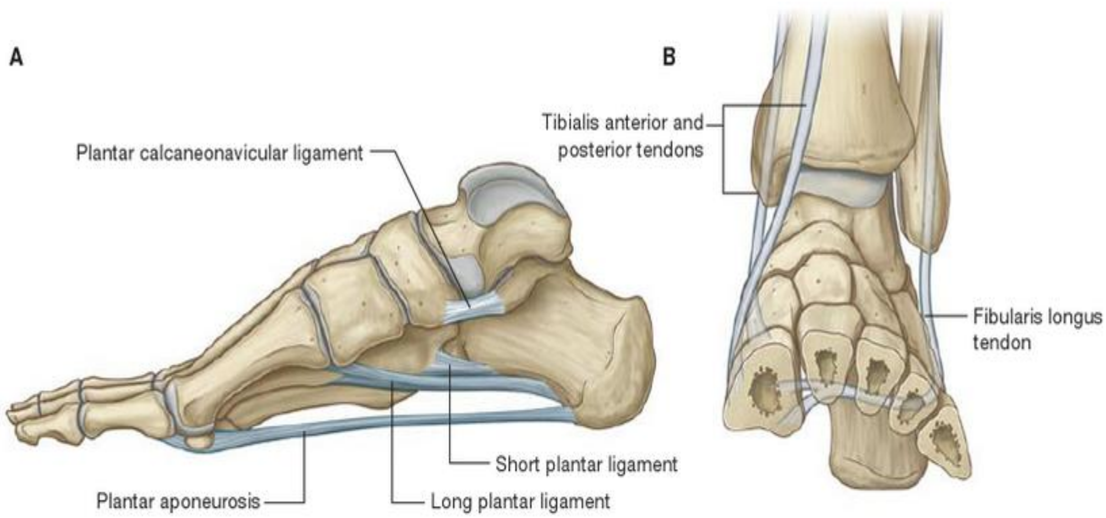

Figure 2.1. Support for the arch of the foot with A displaying a medial view of the longitudinal arch along with the ligaments supporting it and B displaying the transverse arch along with the ligaments and muscle tendons supporting it (Drake, R., Vogl, A. W., & Mitchell, A. W. M. (2009). *Gray's Anatomy: Gray's Anatomy for Students (2nd Edition)*. Saint Louis, MO, USA: Elsevier - Health Sciences Division. Retrieved from http://www.ebrary.com; p. 614, Fig. 6.108).

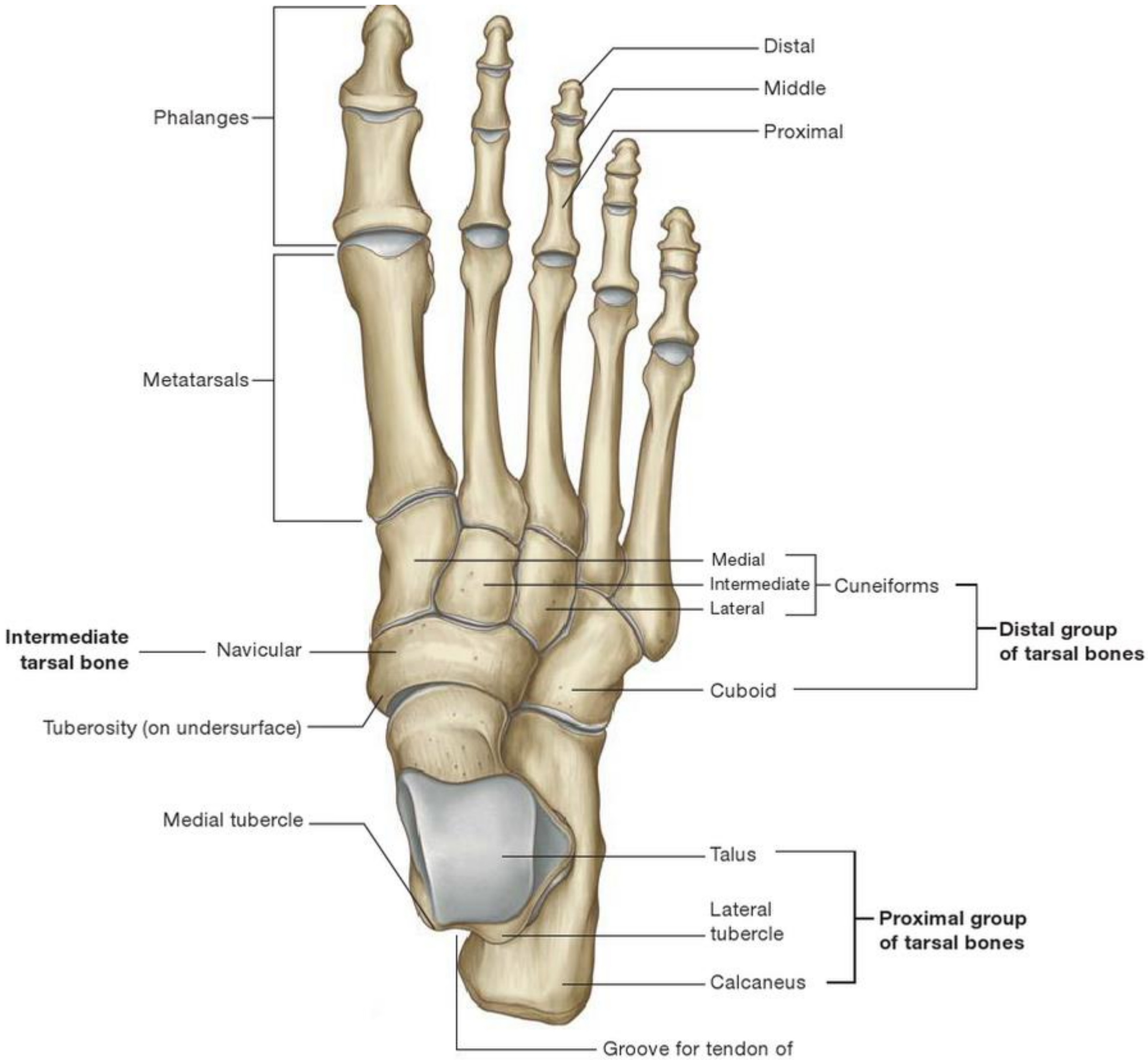

Figure 2.2. The bones of the foot from a transverse view, as well as their division into the distal tarsal bones, proximal tarsal bones, metatarsals and phalanges, (Drake, R., Vogl, A. W., & Mitchell, A. W. M. (2009). *Gray's Anatomy: Gray's Anatomy for Students (2nd Edition)*. Saint Louis, MO, USA: Elsevier - Health Sciences Division. Retrieved from http://www.ebrary.com; p. 601, Fig. 6.91).

The foot is a structure made up of many of bones, generally divided into tarsal bones, metatarsals and phalanges (Figure 2.2). There are seven bones that make up the tarsal bones, organized into a distal and proximal row, with the navicular bone between the two rows on the medial side (Drake, 2009). Most inversion and eversion occur at the tarsal bones. From the tarsal bones, the foot branches out into individual toes, made up of the metatarsals and phalanges, with the two being connected by the metatarsophalangeal joints. The tarsometatarsals joint, connecting the tarsals to the metatarsals, allow limited sliding movement, where the metatarsophalangeal joint allows flexion, extension, adduction and abduction of the foot (Drake, 2009). Independent movement of the metatarsals is limited by the transverse metatarsal ligaments. The metatarsals and tarsal, with the help of muscles and ligaments, are orientated in a way that forms the longitudinal and transverse arches, which acts to absorb and transfer forces during walking and standing (Drake, 2009). Both of the arches end with the metatarsals, however, the longitudinal arch starts at the calcaneus, where the transverse arch begins at the talus.

The arches of the foot are supported by a variety of ligaments and muscles. The main ligaments involved in supporting the arch of the foot (Figure 2.1) include the plantar calcaneonavicular, plantar calcaneocuboid, plantar aponeurosis and long plantar ligaments (Drake 2009). The muscles that provide dynamic support for the arch while walking include the tibalis anterior and posterior and the fibularis longus (Figure 2.1). The anatomy of the foot is quite complicated for 3D modelling during shod walking, which is why it is not uncommon in shod walking to model the foot as a rigid body.

2.1.2 Ankle

The ankle is composed of multiple joints (Procter and Paul, 1982). The ankle joint allows the weight of the body to be transferred from the fibula to the talus (Martini et al., 2003). Procter and Paul (1982), describe the upper ankle joint as primarily responsible for flexion (dorsiflexion)/extension (plantar flexion), where the lower ankle allows the inversion/eversion of the hind foot relative to the talus. The joint allowing for flexion/extension is composed of the tibia, fibula and talus bone of the foot. The malleolus of the tibia and fibula form a deep socket around the inferior surface of the tibia, allowing for the talus to sit securely in the socket while allowing flexion/extension (Drake et al., 2009). The ankle is supported laterally by the medial ligament (Drake et al., 2009) and laterally by the anterior talofibular ligament, the posterior talofibular ligament and the calcaneofibular ligament (Cooper, 2008). Ligaments are designed to resist force in one direction and are meant to carry loads in the direction that the individual fibres run (White and Panjabi, 1990). The subtalar joint consists of the articulations between the talus and calcaneus bones, and is the joint allowing inversion and eversion of the foot (Cooper, 2008). The subtalar joint is supported by the lateral, medial, posterior and interosseous talocalcaneal ligaments. While the ankle is really made up of multiple single axis joints, it is commonly simplified to a single flexion/extension and inversion/eversion axis. Pronation for example is a combination of dorsiflexion and eversion, while the bones of the foot are being abducted (Manter, 1941). For the purpose of this thesis, the ankle will be presented as having a flexion/extension, inversion/eversion and an internal/external rotation axis.

There are many foot muscles that originate on the leg, and therefore cause ankle movement. The superficial muscles are the primary movers of the ankle joint during locomotion (Martini et al., 2003). The tibialis anterior is the primary muscle responsible for dorsiflexion, and also generates inversion of the ankle joint (Martini et al., 2003). Similarly, both heads of the gastrocnemius and tibialis posterior are also active during inversion, however they primarily act as plantar flexors with the soleus muscle. The peroneus muscles are the primary muscles responsible for eversion and are also active during plantar flexion (Martini et al., 2003). While they act during ankle motion, the peroneus longus, tibialis anterior and tibialis posterior insert on the undersurface of the bones of the foot and act to support the arch of the foot (Drake et al., 2009) (Figure 2.1). Gastrocnemius on the other hand is the only one of the muscles to cross the knee, acting as a knee flexor as well as an ankle mover via the Achilles tendon. Similar to the muscle of the knee, hip and pelvis, these muscles act to stabilize the ankle, as well as move it.

2.1.3 Knee

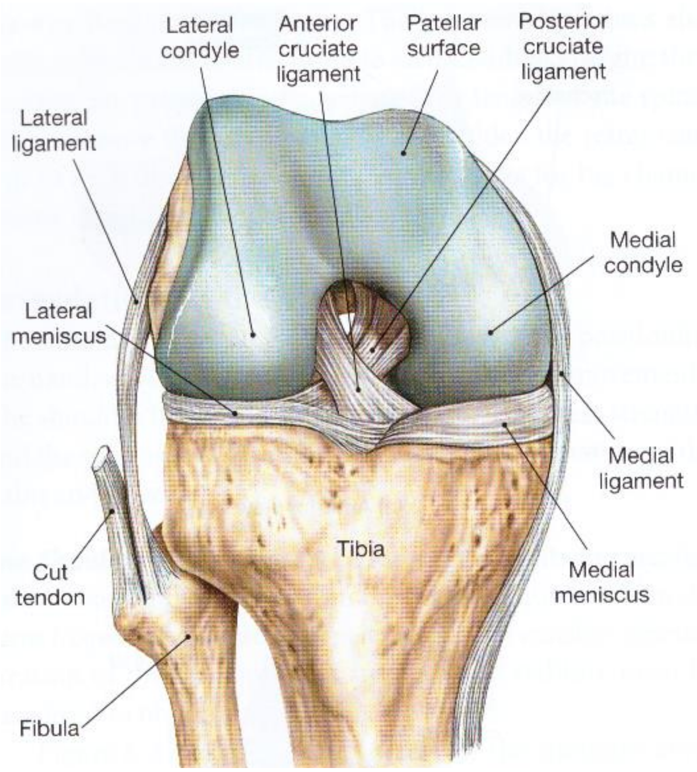

Figure 2.3. Represent the anterior view of the knee and is supporting soft tissue. The condyles of the femur are represented by the titanium colour. (Martini, F. H., Bartholomew, E. F., Ober, W. C., Garrison, C. W., Welch, K., & Ralph. Hutchings. (2003). *Essentials of anatomy & physiology 3$_{rd}$* Edition. Upper Saddle River: Pearson Education, Inc.; p. 154, Fig. 6-26).

The knee is a combination of three separate joints, allowing for the movement and for the weight of the femur to be transferred to the tibia (Martini et al., 2003). Two of the joints connect the lateral and medial condyles of the femur to the tibia (Figure 2.3). The last joint is between the patella and the femur (Martini et al., 2003). While the primary motion of the knee is flexion/extension, some movement in the transverse plane does occur (Drake et al., 2009). A pair of fibrocartilage pads lie between the medial and later condyles of the femur and tibia. Called menisci, these pads act as a cushion and conform to the shape of the articulating surface

(Martini et al., 2003). The knee has four major ligaments that provide stability to the joint. The two collateral ligaments run along the medial and lateral sides of the joint, stabilizing the hinge like motion during flexion and extension (Drake et al., 2009). The cruciate ligaments limit the amount of anterior and posterior movement of the knee (Martini et al., 2003). Both cruciate ligaments originate on the wall of the intercondyle fossa of the knee, with the posterior ligament inserting on the posterior intercondylar area of the tibia, limiting posterior movement of the tibia. The anterior ligament inserts on the anterior intercondylar area of the tibia, limiting anterior movement of the tibia (Drake et al., 2009). In addition to the musculature surrounding the knee, passive tissues help stabilize the knee during motion.

The muscles of the knee that are found on the anterior and lateral aspect of the limbs extend the joint, and muscles found on the back of the limbs flex the joint (Martini et al., 2003). The quadriceps muscles make up the main knee extensors. While the quadriceps attach to the tibial tuberosity via the patella and patellar ligament, the rectus femoris' origin crosses the hip joint, with the three vastus muscles originating on the femur. The three hamstring muscles run along the posterior side of the knee and make up the primary knee flexors (Martini et al., 2003). It is made up of the bicep femoris, semitendinosus and semimembranosus muscles, all of which originate on the posterior pelvis. In addition to its origin on the pelvis, the bicep femoris has a short head, origination on the posterior femur (Drake et al., 2009). These muscles and ligaments are designed for their own function, but a combination of forces from these muscles is required to stabilize the knee during the stance phase of bipedal locomotion (Shelburne et al., 2005).

2.1.4 Pelvis

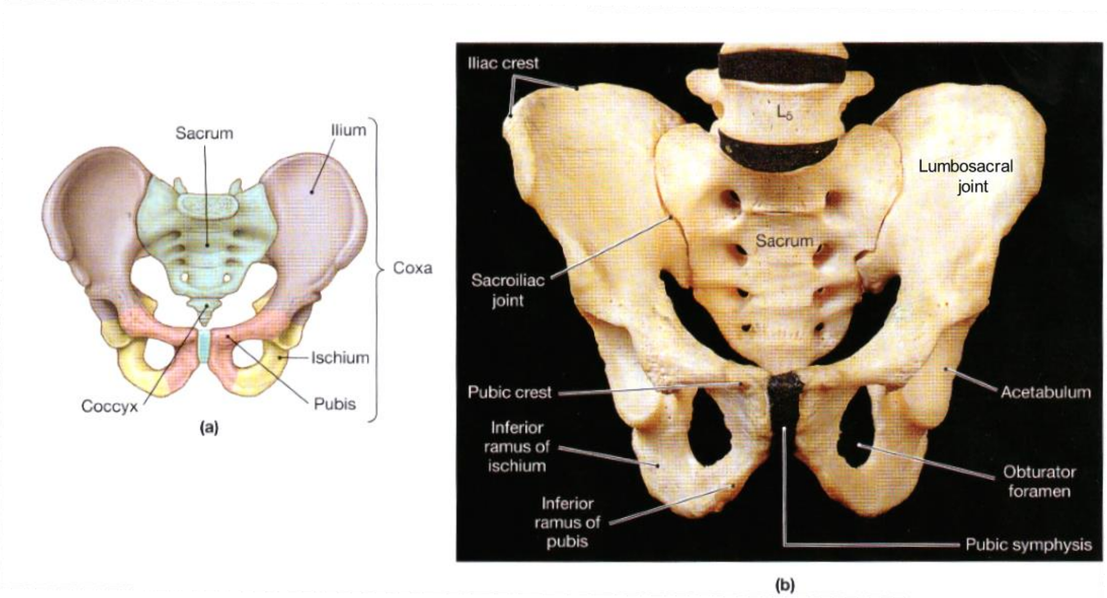

Figure 2.4. Displays the anterior view of the pelvis. (a). Illustrating the bones of the pelvis. (b). Anterior view of the bony landmarks and joints of the pelvis, as well as the connection to lumbar spine. (Modifed from Martini, F. H., Bartholomew, E. F., Ober, W. C., Garrison, C. W., Welch, K., & Ralph. Hutchings. (2003). *Essentials of anatomy & physiology 3rd* Edition. Upper Saddle River: Pearson Education, Inc.; p. 154, Fig. 6-26).

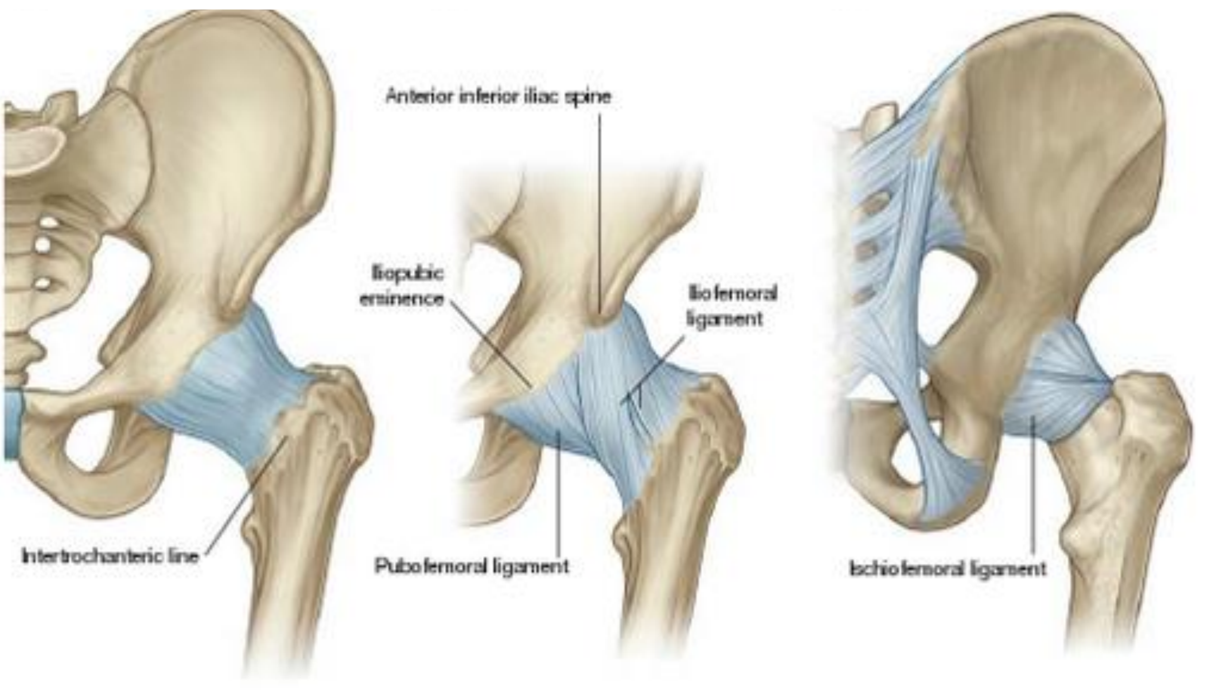

Figure 2.5. Represents the fibrous membrane and ligaments of the hip joint. A. Anterior view of fibrous membrane. B. Anterior view of illiofemoral and pubofemoral ligamens. C. Posterior view of ischiofemoral ligament (Drake, R., Vogl, A. W., & Mitchell, A. W. M. (2009). *Gray's Anatomy: Gray's Anatomy for Students (2nd Edition)*. Saint Louis, MO, USA: Elsevier - Health Sciences Division. Retrieved from http://www.ebrary.com; p. 535, Fig. 6.32).

The pelvis is made of fused bones, allowing for movement of the legs at the hip, and the spine at the lumbosacral joint, as well as the transfer of load from the upper body to the lower limbs and eventually to the ground (Drake et al., 2009). The pelvis is comprised of two large hip bones (coxa), the sacrum and the coccyx (Figure 2.4). The hip bones are comprised of the ilium, ischium and the pubis. The coxa are joined posteriorly by the sacrum, at the sacroiliac joint, and anteriorly by the pubic symphysis (Martini et al., 2003). The lumbosacral joint is formed between the L_5 and S_1, and contains the intervertebral disc that joins the two vertebral bodies (Drake et al., 2009). This in theory allows for movement in all directions, however due to constraints on the disc and vertebral geometry, ≈25° of the range of motion commonly attributed to the spine is due to movement at the hip (White and Panjabi, 1990). The stability of the hip joint is reinforced by strong iliolumbar and lumbosacral ligament, which run from the illum and sacrum, to the L_5 vertebrae, in a variety of directions. With the help of the ligaments present in Figure 2.5 the pelvis connects to the hip, where a ball and socket joint is formed between the hip bones acetabulum and the head of the femur (Martini et al., 2003), allowing the femur to have six degrees of freedom.

The detail of the hip joint and pelvis is paramount to understanding spine kinematics, as some of the muscles attaching to the pelvis help move the spine (discussed in 2.1.5), while some assist in lower limb movement. Gluteal muscles run along the lateral and posterior aspects of the femur to the pelvis. While the gluteus maximus produces extension and lateral rotation at the hip, the gluteus minimus and gluteus medius cause abduction and medial rotation (Martini et al., 2003). Hip adductors run along the inside of the thigh, inserting on the pelvis, causing adduction, medial rotation and flexion (Martini et al., 2003). Major flexors of the hip, such as iliopsoas, originate on posterior abdominal wall and descend through gaps in the pelvis, attaching to the femur (Drake et al., 2009). In addition to causing movement of the hip, these muscles control the movement of the pelvis during single leg weight bearing (Drake et al., 2009)

2.1.5 Vertebral Column

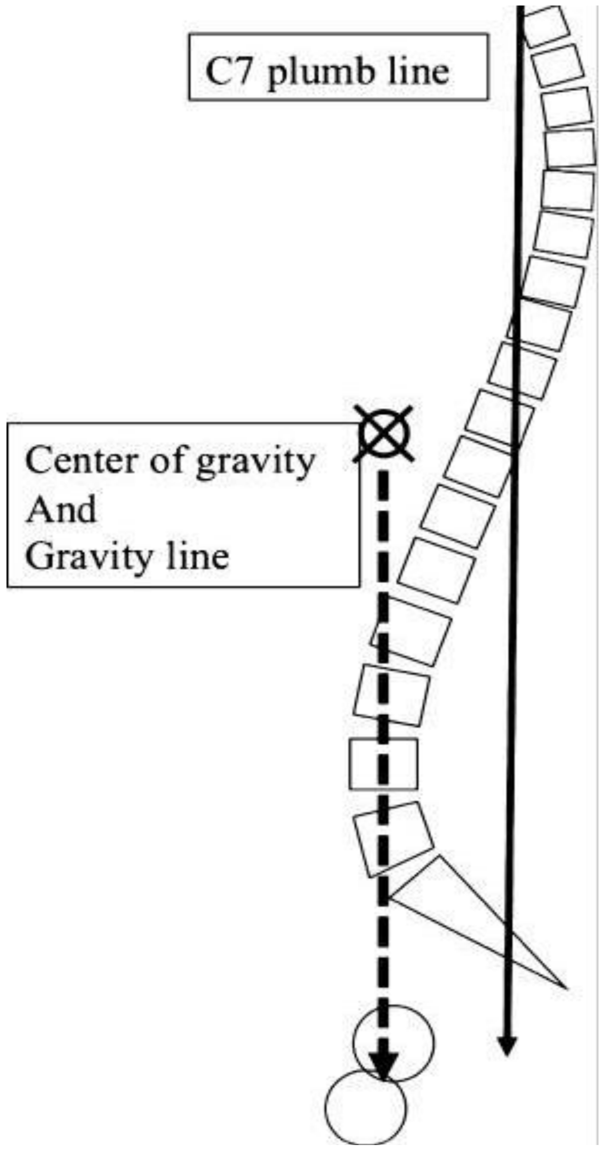

Figure 2.6. Illustrates the vertebral column, with the anterior aspect of the spine facing left. The centre of gravity line for the upper body passes in front of the spine. The C7 plumb line is an imaginary vertical line originating in the centre of C7 (Roussouly, P., & Pinheiro-Franco, J. L. (2011). Sagittal parameters of the spine: biomechanical approach. *European Spine Journal*, *20*(5), 582. Fig. 6 with kind permission from Springer Science and Business Media)

The human spines' bony structure is comprised of 26 vertebrae, making up the vertebral column (Martini et al., 2003). The 26 vertebrae are divided into four regions: seven cervical, 12 thoracic, five lumbar, one sacrum and one coccyx (Tortora, 2005). The sacrum and coccyx are made up of multiple fused vertebrae, with the coccyx being made of three-four fused vertebra and the sacrum being made up of five (Tortora, 2005). The cervical and lumbar regions display curves that are convex anteriorly, where the thoracic and sacral regions display curves that are convex posteriorly (White and Panjabi, 1990). Due to the curvature of the spine, the body's centre of gravity only pass through the regions of the spine that are anteriorly convex, with the cervical spine supporting the head, and the lumbar spine supporting the weight of the upper body (Figure 2.6) (Bassett, 2005). Most of the vertebral bodies do not directly articulate with one another, as they are separated by an intervertebral disc (IVD). There are no intervertebral discs found between the sacrum and the coccyx, or between the first and second cervical vertebrae (Martini et al., 2003). Two connected vertebrae, as well as the intervertebral disc and ligaments that connect them, are defined as a functional spine unit (FSU) (White and Panajbi, 1990).

2.1.6 Intervertebral Disc

The IVD is found between the cartilaginous end-plates of the inferior and superior vertebral body and is composed of the nucleus pulposus and the annulus fibrosus (White and Panjabi, 1990). The nucleus pulposus is a soft, elastic, semi-fluid mass, which allows the IVD to compress and distort while loaded (Martini et al., 2003; Tampier et al., 2006). The nucleus pulposus is contained superiorly and inferiorly by the end-plates, as well as a layer of fibrocartilage, called the annulus fibrosus, composing the lateral outer boundary (Martini et al., 2003). The annulus fibrosus is divided into concentric laminated bands. Adjoining bands are angled at approximately 30°, but run opposite in direction, making angle of 120° between them

(White and Panjabi, 1990). The annulus fibrosus is anchored to the cartilaginous end-plates on the inner layer, and the vertebral body on the outer layer. The cartilaginous end-plate is composed of hyaline cartilage, and bulges up into the vertebrae when the spine is loaded (compressed). While all the discs are designed to absorb shock and distribute force, some vertebrae support more of the body then others.

2.1.7 Ligaments of the Functional Spine unit

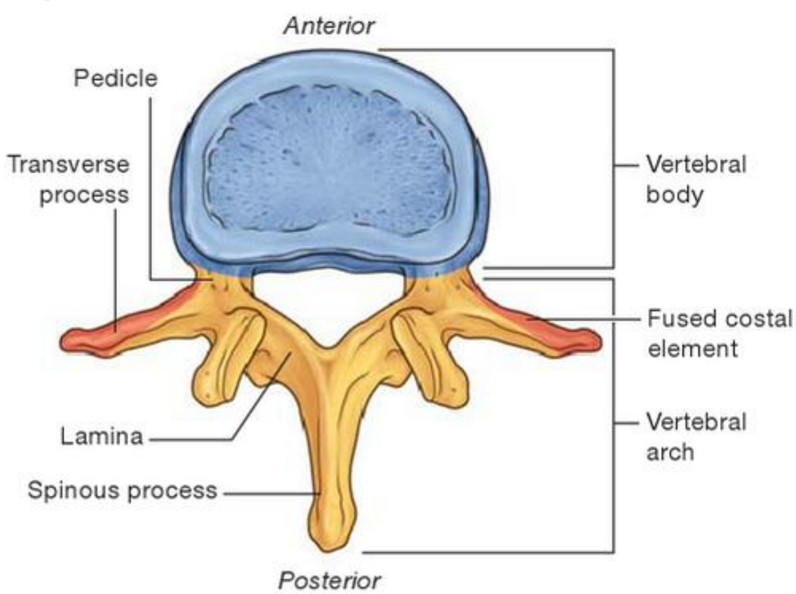

Figure 2.7. Represents a superior view of a typical vertebrae. (Modified from Drake, R., Vogl, A. W., & Mitchell, A. W. M. (2009). *Gray's Anatomy: Gray's Anatomy for Students (2nd Edition)*. Saint Louis, MO, USA: Elsevier - Health Sciences Division. Retrieved from http://www.ebrary.com; p. 60, Fig. 2.6).

There are six main ligaments that run between the vertebrae of the cervical, thoracic and lumbar spine. The posterior longitudinal, ligament flava, capsular, interspinous and supraspinous ligaments, run along the posterior aspect of the vertebral bodies (White and Panjabi, 1990), where the anterior longitudinal ligament runs along the anterior aspect of the spine (Drake et al., 2009). The thoracic region has additional ligaments, the intertransverse ligaments, connecting each of the transverse process (Figure 2.7) to the articulating vertebrae below. The ligaments of the upper cervical spine (occiput to C2) are quite different than the rest of the spine (White and Panjabi, 1990) and will not be examined in this literature review. The anterior and posterior longitudinal ligaments, as well as the supraspinous ligament, run the entire length of the spine connecting to each vertebrae. The anterior longitudinal ligament inserts at the base of the skull and extends inferiorly, originating on the sacrum, connecting to the anterior aspect of each vertebral body and intervertebral disc along the way (Drake et al., 2009). Similarly, the posterior longitudinal ligaments insert at the skull, connecting to the posterior aspect of each vertebral body and intervertebral disc along the way, origination on the coccyx (White and Panjabi, 1990). The supraspinous ligament runs from the C_7 vertebrae to the sacrum, connecting to the tips of each vertebral spinous process. The supraspinous ligament merges with the ligamentum nuchae in the cervical spine (Drake et al., 2009). The interspinous ligament connects to adjacent vertebral spinous process and blends with the supraspinous, where the ligament flava exist on each side of a vertebrae, attaching to the laminae of the adjacent vertebra (Drake et al., 2009). Similarly the capsular ligaments connect two adjacent vertebra, connecting just behind the margin of the articular process. The ligament flava is made up of mostly elastic fibres, and represents the most pure elastic tissue in the body (White and Panjabi, 1990).

Ligaments act similar to elastics, resisting tensile forces and typically buckling under compression (White and Panjabi, 1990). They also act to restrict the separation of two adjacent vertebrae on the same side of the spine as the ligament.   In the spine, ligaments are reported to have four main functions: reduce the energy of the musculature during stabilization, protect the spinal cord by restricting motion, provide stability to the spine within its range of motion, and protect the spinal cord during traumatic situations (White and Panjabi, 1990). The anterior and posterior longitudinal ligaments, in addition to resisting the separation of vertebrae, resist the bulging of the intervertebral disc due to compression (White and Panjabi, 1990). While all of these ligaments stabilize the spine, without the assistance of muscle, they do not act to fully protect it from deformation. This was documented by Lucas and Bresler (1961), who observed the spine buckling under a load of 2 kg when isolated from the muscles.

2.1.8 Spine Musculature

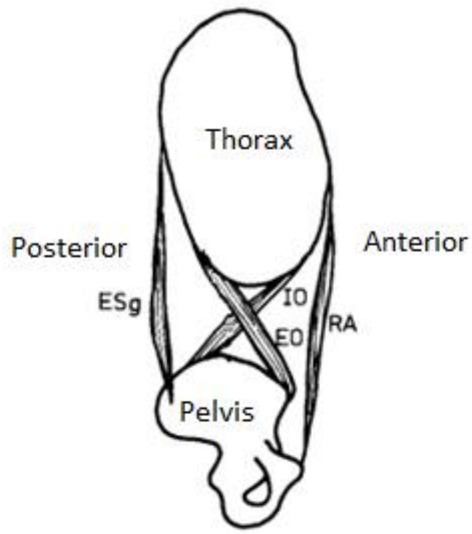

Figure 2.8. Represents a sagittal view of the global muscle system, running from the thorax to the pelvis (Modified from Bergmark, A. (1989). Stability of the lumbar spine: a study in mechanical engineering. Acta Orthopaedica, 60(S230), 1-54; p. 20, Fig. 5-2 www.tandfonline.com)

The muscles of the abdomen and back are required to stabilize the spine, similar to the use of guy wires supporting a radio tower (Bergmark, 1989). McGill (2007) continued to describe the role of muscles as guy wires, discussing their critical role in ensuring spine stability during loading, as well as maintaining postures. Bergmark (1989) divides the stabilizing musculature into the local and global system. Muscles of the local system have separate origin and insertion for each functional spine unit, and are used to maintain curvature and give sagittal and lateral stiffness to the spine. Drake et al. (2009), refers to these muscles as transversospinales muscles, and consist of such muscles as; semispinalis and multifidus. These muscles are on the posterior aspect of the spine and when acting together cause extension. Only one side of these muscles can also be activated, causing lateral bending towards that side (Drake e al., 2009). The global system consists of active components and transfers the load between the thoracic cage and the pelvis (Figure 2.8) (Bergmark, 1989). Bergmark considers this group to be made up of the erector spinae muscles, the internal and external obliques, the rectus abdominus muscles, and the lateral quadratus lumborum. It should be noted that the erector spinae muscle group has both global and local aspects (Bergmark, 1989). The thoracic part composes the majority of the muscle and belongs to the global system, where the lumbar erector spinae supports the local system. All of the global muscles run from the thoracic cage to the pelvis (Bergmark, 1989). The output of these active muscles is based on the external load and the muscle length (White and Panjabi, 1990). Considering the link between external load and low back pain, the load on the spine and the muscles supporting it are major factors in predicting spinal load severity. Due to the lack of direct measurement regarding the internal load, a variety of indirect methods are used, with techniques such as comparing EMG to force being typical (White and Panjabi, 1990).

Each muscle of the global system has a different purpose in spine movement The erector spinae muscle group acts to extend the spine and are primarily used to maintain an erect spine (Martini et al., 2003). Moving laterally from the spine, Martini et al. (2003) subdivides this group into the spinalis, longissimus and iliocostalis division. The longissimus and iliocostalis can be further subdivided into pars lumborum and the pars thoracis. The pars lumborum division of these two muscles generates posterior shear forces that attempt to counter the anterior shear forces produced during certain movements, such as flexion (McGill, 2007). The pars thoracis of the iliocostalis and the longissimus have greater moment arms then the pars lumbar counterparts, and are the primary back extensor (McGill, 2007). These muscle activated bilaterally cause spine extension, or unilaterally to cause lateral bending of the thoracic cage to the active side of the spine. The latissimus dorsi muscles runs along the back, originating on the lumbar spine and inserting on the humorous (Drake et al., 2009). While this muscle may be indirectly influencing the lumbar spine, it is not considered to have a sustained role in maintaining spine stability (Bergmark et al., 1989). The oblique muscles and the rectus abdominus muscles are considered to be axial muscles of the trunk, running from the start of the thoracic spine to the pelvis (Martini et al., 2003). The internal oblique fibres run along the side of the body, originating along the iliac crest and inserting on the cartilage of the ribs. Their main function, when activated bilaterally, is to cause spinal flexion, where unilateral activation causes side bending and rotation of the thoracic cage (Bergmark, 1989). The functional difference between the internal and external oblique is that the external oblique causes the chest to rotate towards the opposite side, where the internal oblique causes it to rotate to the same side (Drake et al., 2009). The rectus abdominus muscles are the most important flexors of the spinal column, opposing the erector spinae (Martini et al., 2003). Although it may not be their primary function, the muscles of the torso act to support the spine against external loads.

2.1.9 Anatomical Literature Review Summary

The physical structures of the spine and the lower limbs form the kinematic chain and may contribute to the development of low back pain. While the weight of the body is passed through joints from bone to bone, muscles and ligaments of the body work together to provide joint stability. Winter (1980), discusses how the collaboration of muscles at all three lower limb joints is required to prevent collapsing during weight bearing. Spinal stability requires the stabilizing muscles to pull on the spine in opposite directions with an equal tension, similar to guy wires on a radio tower, preventing it from buckling in any one direction (McGill, 2007).

The elastic like structure of ligaments, connects different joints, and prevents their separation (Martini et al., 2003). This causes the pulling on one bone as a result of the movement of another. In addition to stabilizing the vertebral column, ligaments and muscles of the spine cause compression to the intervertebral disc when matching the external load experienced by the body (White and Panjabi, 1990). Although there is no direct way to measure the exact load experienced by joints without disruption to the joint (i.e. insertion of a transducer), the activation of the supporting muscles can be used to provide an estimate (Bergmark, 1989), or indication of joint loading. While the anatomy of the foot was not fully described in this review of the literature, it would also play a role in dictating movement at the ankle. The anatomy of the spine, pelvis and lower limbs, helps to explain the theory of the kinematic chain and how movement at one joint can result in movement at another.

2.2 Stance Phase Movement Literature Review

The gait cycle is made of four phases, two single support and double support phases (Winter, 2009; Abboud, 2002). Each single support phase makes up approximately 40-45% of the gait cycle, with the remaining 10-20% being considered double support (Winter, 2009; Abboud, 2002). A limb's stance phase is the entire time a limb is weight bearing, from initial contact (also called heel strike) to toe off (Tiberio, 1987; Abboud, 2002). Most of the disorders associated with the foot are related to the weight-bearing process of the stance phase (Abboud, 2002).

2.2.1 Ankle

The ankle joint experiences movement in multiple axes during stance. The heel strikes the ground dorsiflexed, and plantar flexes immediately following contact (Abboud, 2002). This movement ranges from ≈5° plantar flexion, to ≈10° dorsiflexion (Kadaba et al., 1990). At heel strike the ankle is also slightly supinated (Tiberio, 1987). Following heel strike the ankle begins to pronate, reaching maximum pronation when the entire foot is on the ground (Tiberio, 1987). The motion is then reversed, with the ankle supinating until the foot is off the ground. This entire process ranges from approximately ≈5° supination, to ≈5° pronation (Tiberio, 1987). Closed chain pronation, where the foot is weight bearing and fixed on the ground is believed to transfer up the ankle into internal rotation of the tibia. This is a result of the calcaneus everting during pronation, which causes the talus to rotate medially due to the moment created by the pulling force produced by the ligaments connecting the two bones (Tiberio, 1987). Due to the tight fit of the talus in the socket created by the tibia and fibula, the ankle joint forces the lower limb to internally rotate (Figure 2.9). It is this aspect of internal rotation, as a result of pronation, that insoles often attempt to correct for (Lafortune et al., 1994; Eng and Pierrynowski, 1994; Bird and Payne, 1999; Marinakis and Catalfamo, 2004). Bird and Payne (1999) discussed how this is the basis behind the theory of the kinematic chain. With internal rotation of the tibia theoretically causing internal rotation of the leg and anterior pelvic tilt (Bird and Payne, 1999). While they do appear to be linked, Reischl et al. (1999), found no link between peak foot pronation and peak tibial rotation.

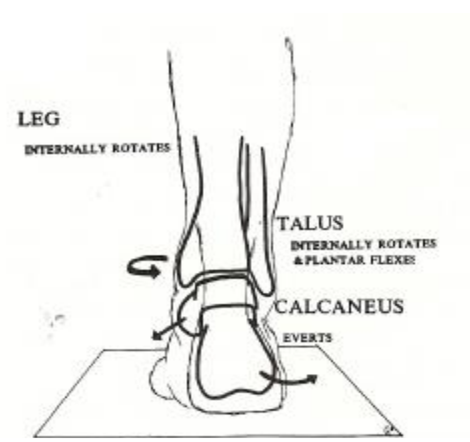

Figure 2.9. Represent the posterior view of the lower leg and foot. Illustrates how pronation of the foot leads to internal rotation of the tibia. (Modified from Tiberio, D. (1987). The effect of excessive subtalar joint pronation on patellofemoral mechanics: a theoretical model. *Journal of orthopaedic & Sports physical Therapy*, *9*(4); p. 162 Fig.3).

2.2.2 Knee

The knee has mostly uniaxial movement about the flexion/extension axis during the stance phase (Kozanek et al., 2009). The knee is almost fully extended during the start of stance phase and begins to flex following heel strike, and similar to the motion at the ankle, continues to flex until the entire foot is in contact with the ground (Tiberio, 1987). Following the maximum flexion point, the knee begins to extend, approximately reaching its initial angle during heel contact prior to toe off (Tiberio, 1987; Kadaba et al., 1990). This range of knee flexion was reported to be ≈0° at heel strike, reaching a peak between ≈8-13° (Tiberio, 1987; Kadaba et al., 1990; Kozanek et al., 2009). There is slight motion about the frontal and transverse plane. Knee flexion was accompanied by internal rotation, with the external rotation occurring as the knee extended (Tiberio, 1987; Kozanek et al., 2009). This rotation movement was reported by Kozanek et al. (2009) to be between ≈1.5° internal rotation and ≈7.4° external rotation. Using fluoroscopic imaging, Kozanek et al. (2009), reported a similar movement for frontal plan motion, with valgus accompanying flexion and varus accompanying extension. The range of this motion was reported to have a minimum of ≈3.2° valgus at heel strike, with valgus peaking at ≈5.7°. Using a passive-reflective motion capture system, Kabada et al. (1990), reported a similar trend between valgus/varus and flexion/extension, with angles shifted a few degrees towards varus.

2.2.3 Hip

Similar to the knee and ankle, the hip displays motion primarily in the sagittal plane during walking. At heel strike the hip appears to be in a flexed position, which plateaus as the hip accepts weight. Following this the hip begins to be in almost constant extension until just prior to toe off (Isacson, et al., 1986; Kabada et al., 1990). This ranges from 25-35° flexion at heel strike and extends until approximately 1-5° past neutral (Isacson, et al., 1986; Kabada et al., 1990). While the hip is in its flexed position following heel strike, it also appears to be externally rotated and almost neutral in the frontal plane. Hip frontal plane movement appears to follow sagittal plane movement, starting from neutral, hip flexion is accompanied by adduction and extension by abduction (Kabada et al., 1990). This ranges from ≈5° adduction to ≈5° abduction. In terms of rotation, the hip is slightly internally rotated at heel strike, and almost continuously internally rotates until toe off (Isacson, et al.; 1986, Kabada et al., 1990). This was documented ranging from ≈5° internal rotation, to ≈0° external rotation prior to toe off. This continuous internal rotation is only partially a result of the internal rotation of the femur accompanying heel strike (Reischl et al., 1999). While there was a relation between the timing of peak tibial rotation and peak femur rotation, there was no relation between the timing of peak femoral rotation and peak pronation (Reischl et al., 1999).

2.2.4 Pelvis

Motion in the pelvis during the stance phase occurs in all three planes. Sagittal movement of the pelvis appears to oscillate around the angle experienced during either foots' heel strike. The pelvis shows very little movement in the sagittal plane, increasing and deceasing posterior tilt around this point by 1-2° (Thurston and Harris, 1983; Vink and Karseem, 1988; Kabada et al., 1990). Conversely, in the transverse plane, the pelvis is at its most extreme position following heel strike (Thurston and Harris, 1983; Kabada et al., 1990). The pelvis is twisted towards the trailing foot by approximately ≈3° following heel strike, and almost constantly moves towards being ≈3° twisted in the other direction prior to toe-off. (Thurston and Harris, 1983; Kabada et al., 1990). Motion in the frontal plane appeared to follow a similar pattern, with the most extreme angle however occurring at toe off. The pelvis is tilted laterally between 5-10°, towards the swinging limb following toe off, it then tilts in the other direction, reaching 5-10° by the time the swinging foot makes contact with the ground (Thurston and Harris, 1983; Kabada et al., 1990). The pelvis appears to be in a neutral position in the frontal plane during the middle of the double support phase. (Thurston and Harris, 1983; Kabada et al., 1990)

2.2.5 Spine

The motion of the lumbar spine and trunk appears to follow the opposite movement pattern of the pelvis (Thurston and Harris, 1983; Callaghan et al., 1999). Sagittal plane lumbar spine movement appeared to oscillate around a point in mid swing phase (Thurston and Harris, 1983), increasing and decreasing by approximately 3° around this point, with maximum flexion around toe off and maximum extension occurring around heel strike (Thurston and Harris, 1983; Callaghan et al., 1999). Considering left stance phase in the frontal plane, maximum lumbar bending to the left occurred at the start of left single support phase and was ≈10° (Thurston and Harris, 1983; Callaghan et al., 1999). Continuing the left step example, after the spine reaches maximum lateral bend it, it begins shifting to the right, reaching a maximum of similar magnitude as right single support phase begins (Rowe and White 1996; Callaghan et al., 1999). In the transverse plane, the spine oscillated left and right by ≈3-6° (Thurston and Harris, 1983; Rowe and White, 1996). Between heel strikes the spine twists at a relatively consistent rate towards the lead foot (Callaghan et al., 1999). A large portion of the spine's relative motion during walking is a result of the pelvis, as the relative angle of the spine is partially based on the pelvis (Callaghan et al., 1999). While some studies have examined the relative motion of the entire trunk (Callaghan et al., 1999) and the lumbar spine (Thurston and Harris, 1983; Rowe and White, 1996), no study was found examining the angle of the thoracic spine, relative to the lumbar spine, during gait. Considering the relative similarity between the reported range of motion examining the entire spine during walking, and those examining just the lumbar spine, it is expected that thoracic spine motion will be similar to the movement patterns of the trunk and lumbar spine.

2.2.6 Movement Pattern Literature Review Summary

Movement of the lower limbs during gait is primarily in the sagittal plane, where movement of the spine and pelvis appears to occur more equally in all three planes. The knee and hip internal rotation that does occur is believed to be linked to pronation of the foot (Lafortune et al., 1994; Eng and Pierrynowski, 1994; Bird and Payne, 1999; Marinakis and Catalfamo, 2004). However, Reischl et al. (1999) reported that there was no relationship between the magnitude of peak pronation, and peak internal rotation of the tibia or femur. Pelvis motion experiences its maximum frontal plane motion around toe off, in preparation of one limb swinging, while motion in the sagittal plane reaches maximum at heel strike, as one foot stretches forward (Thurston and Harris, 1983; Kabada et al., 1990). The motion of the pelvis is also partially responsible for the movement observed in the relative lumbar spine and trunk angles (Callaghan et al., 1999).

2.3 General Methodology Literature Review

2.3.1 Kinematic Motion Capture

When observing complex 3D movement an imaging system is the best method available for capturing all the data required for kinematic analysis (Winter, 2009). This is by comparison to other direct measurement techniques, such as goniometers and accelerometers. For this reason, a three dimensional passive optoelectronic motion capture system was used for this collection, specifically a seven camera Vicon MX system (Vicon Systems Ltd., Oxford, UK).

Markers are typically placed over anatomical landmarks, allowing for the reconstruction of limbs as segments. These segments are really an estimate of limb motion, as they are not the same as the movement of the bony structures they represent (Cappozzo et al., 1995). Once segments are created the coordinate data of the markers are used to obtain the global position of that segment in space (Cappozzo et al., 1997; Winter, 2009). In order to obtain the global position of a segment, multiple cameras must detect a minimum of three markers assigned to the segment (Cappozzo et al., 1997; Vicon Motion Systems Ltd.). Once multiple segments are established in a global reference system, relative angles can be created from the global angles of two segments (Winter, 2009).

While passive reflective motion capture systems are extremely accurate, they are not without error. In addition to the error associated with skin-mounted marker positions not being the exact location of the underlying bone or bones within the segment, the markers' location relative to their anatomical landmark changes with skin movement. This is known as skin artifact and can be a major source of error (Cappozzo, 1991). Cappozzo (1991) also discussed the idea of instrumental error, which can be subdivided into systematic and random error. Systematic error is due to errors associated with calibration and random error is a result of quantization problems in the image and digitizing processing (Cappozzo et al., 1991). These are errors associated with passive marker based optoelectronic motion capture, however when accompanied by proper experimental and post processing techniques to reduce the error, they do not outweigh the benefits of having three dimensional kinematics without adding encumbering equipment (Winter, 2009).

Proper anatomical landmarks and filtering techniques must be used to reduce the error during the collection of kinematics. In order to reduce skin artifact during collection, a standard land marking procedure must be used in an attempt to reduce the amount of movement due to the displacement of soft tissue between the marker and the bone (Cappozzo et al., 1996). Studies such as Ensberg et al. (2008), and Mörl & Blickhan (2006), have compared movement of reflective markers placed over spinous processes, to radiographs and magnetic resonance imaging (MRI) and have found reflective markers to be highly correlated with these other imaging techniques. In order to ensure that all three markers for segment construction are over spinous processes, a marker configuration similar to the ones used by Schinkel-Ivy and Drake (2015), can be used. As Cappozzo et al. (1996), describes the problem with some of the lower limb marker placements required for segment generation, proper filter frequency cut-offs must be selected in attempt to reduce this error. Winter (2009) discusses how the fastest moving markers in the study of gait, such as the heel and toe, have power up to a frequency of 6 Hz, with 99.7% of the signals power occurring below his frequency. While there might be some signal above 6 Hz, it has almost entirely the characteristics of noise, and is not the result of the processing of walking itself (Winter, 2009). Therefore through the use of proper experimental protocols and processing methods, these errors can be minimized, thus increasing the fidelity of the collected 3D kinematics.

2.3.2 Electromyography

The electrical signal generated by the depolarization of muscle tissue during a muscle contraction is referred to as electromyogram (EMG), and is briefly described in this section. The membrane potential of a muscle at rest is around -70mV, increasing by nearly 100mV when activated (Lamb and Hobart, 1992). This change in membrane potential causes an action potential that spreads along the entire surface and T-tubules (Martini et al., 2003). This causes the sarcoplasmic reticulum to release stored calcium, which bind to troponin on the actin filament, allowing the formation of myosin cross bridges. Repeated cycles of cross-bridge formation and detachment cause the muscle fibres to shorten (Martini et al., 2003). The electrical impulse that begins this shortening cycle is referred to as an action potential (Martini et al., 2003). The ability for an action potential to occur in a single muscle fibre is controlled by its motor neuron. As a single motor neuron controls anywhere from one to thousands of muscle fibres, all of the muscle fibres controlled by a single motor neuron is referred to as a motor unit (Martini et al., 2003). The signal activating up to thousands of muscle fibres through the recruitment of one motor neuron is called a motor unit action potential, or a MUAP (Winter, 2009). Electrodes placed on the surface of a muscle or inside the muscle (indwelling), will record the algebraic sum of all MUAPs transmitted along the muscle between the electrodes at that point in time (Winter, 2009). Indwelling electrodes may reduce the potential for noise in static exercises, however due to the movement required during walking they were not used in this study. The sum of all MUAPs, or the total muscle force, is dependent on the motor unite recruitment and firing rate (Martini et al., 2003). As the number of motor units is not limitless, normalizing a signal to a reference level of EMG, such as to a maximum voluntary contraction (MVC), is the method of normalization gait that allows for the interpretation of to what degree a muscle is active (Burden et al, 2003). In addition to their activation causing the movement of joints, muscle activation can also be used to determine joint loading (Bergmark, 1989).

2.3.3 General Methodology Literature Review Summary

Three dimensional motion capture and surface EMG can be used to capture kinematics and muscle activation, respectively, during gait. The high correlations between certain surface markers and their anatomical landmarks movement (Mörl & Blickhan, 2006; Ensberg et al., 2008), in combination with filtering techniques (Winter, 2009), can be used to create accurate estimates of a limbs position in space. Clusters can be placed on two adjoining segments to create three dimensional representation of joints during locomotion (Winter, 2009). EMG can be used to determine how much a muscle is active during the recorded kinematics, as EMG displays the sum of motor unit action potentials.

2.4 Previous Insole Research

2.4.1 Insole Categorization

Currently in the marketplace there are numerous types of insoles, each being designed for their own proposed benefits, with most being generally categorized as either hard or soft. Materials such as foam, silicone and natural rubber, are usually used in soft insole construction, as they are designed to absorb shock and cushion the foot in an attempt to alleviate pain (Ball and Afheldt, 2002; Ogon et al., 2001; Shabat et al., 2005). Softer insoles tend to help relieve stress on the foot, but may place additional stress elsewhere in the body. Hard insoles, however, are designed to support the structures of the foot in a specific position (i.e. brace) typically to correct for foot abnormalities and/or pathomechanics (Cambron et al., 2011). A study by Milgrom et al. (1992) on military recruits found that using softer materials resulted in a significant reduction in metatarsal stress fractures and foot overuse injuries, while not reducing the total amount of whole body injuries. This was further supported by Milgrom et al. (2005) that reported no difference in injury rate for recruits using hard, soft or no insole. Furthermore, hard insoles have been suggested to increase the muscles response to movement, causing the muscles of the spine to enhance spine stability (Ogon et al., 2000). Some studies have investigated the effects of customized hard insoles in relation to gait and low back pain; unfortunately these studies are more expensive and require the orthotics to be custom made for each study (Rothbart et al., 1995). There are insoles on the market, such as the ones made by

Barefoot Science Canada Ltd.®, which are adjustable in stiffness, and allow for a more gradual transition to a harder insole, supposedly increasing the strength of the foot. Grouping all the different types of insoles into a hard or soft category is a reflection of the two believed mechanisms behind their use as a LBP treatment.

2.4.2 Questionnaire Based Research

While previous research does credit the use of insoles to prevent LBP as a result of shock absorption or foot realignment, most orthoses and insoles research on LBP is solely questionnaire based. Multiple studies have used questionnaires in combination with insole activity logs to monitor the insoles effect on perceived LBP. Studies by Dananberg and Guiliano (1999), Ferrari (2007), and Cambrom et al. (2011) used the Quebec Back Pain Disability, Oswestry Disability Index, and Visual Analog Scale over six weeks of insole use, and Shabat et al. (2005) used the MILLION questionnaire over five weeks of insole use. The questionnaire results from these four studies indicated that the use of insoles was associated with significant reductions in perceived LBP and perceived lower extremity pain for the LBP population, and increased duration of pain relief, compared to other back pain treatments. Shabat et al. (2005) hypothesized that this reduction in LBP was a result of less fatigue in the back muscles due to the shock absorbing nature of the insoles; however, muscle activation data were not collected to test this assumption. Conversely, studies that have looked at the preventative ability of insoles in a military population have found no significant benefit for LBP. Studies done by Larsen et al. (2002) and Matilla et al. (2010) found no effect of insoles on reducing the incidence of LBP in soldiers. The results of Larson et al. (2002) are inconclusive, as LBP was quantified by the number of days missed due to LBP and soldiers tend to underreport LBP, as most soldiers report no history of back pain in general (Larson et al., 2002). Questionnaire data cannot reveal if insoles alter the biomechanics of tasks such as walking, running or standing, or whether there are other effects.

2.4.3 Kinematic Chain Research

Kinematic studies on insoles typically investigate the kinematic chain. Insoles are commonly designed in a way to place the foot in an altered position, in an attempt to change the orientation of other joints (Lafortune et al., 1994). Proper alignment of the foot is believed to be one of the most crucial functions of shoe insoles and orthotics (Nigg et al., 1999). Segments of the body are connected in a way that they interact much like links in a chain, with pulling at one link in the chain resulting in pulling at another link. This pulling has the potential to pass through several links, resulting in the altered behaviour of other segment. It is believed that this can cause LBP by putting extra strain on the pelvic muscles as a result of increased pronation of the foot's first metatarsophalangeal joint (Bird and Payne, 1999). Effects at the metatarsophalangeal joint in the foot (e.g. eversion/inversion) can cause ankle rotation, affecting tibial rotation, causing misalignment in the knees and patellae, which can then affect the hips and pelvis, finally resulting in an altered flexed posture in the lower back (Bird el al., 2003). Altered posture in the low back is a potential mechanism that causes LBP, as a change in lumbar lordosis results in a change in the angle of pull of the muscles, potentially lowering their ability to resist shear loading (McGill, 2007). Considering that over 50% of a person's mass is comprised of the head, arms and trunk (Winter, 2009), altering trunk position has a large impact on a person's centre of mass and gait kinematics (Saha et al., 2007). Altering ones trunk position can also cause an increase or decrease in EMG. For example, it is possible that an increase in trunk flexion could cause an increase in muscle activation as a result of a larger flexion-bending moment, or a decrease in muscle activation as a result of the trunk now being able to hang of passive tissues without the assistance of muscles (Lühring et al., 2015). This idea of reducing motion not essential to locomotion, in addition to such concepts as reducing impulses and optimizing posture, is one of the primary theories behind the use of insoles to treat LBP.

Multiple studies have examined the effect of insoles on reducing the unnecessary movement associated with pronation. Lafortune et al. (1994) described how it was quite common for insoles to be used in an attempt to reduce the internal tibia rotation that occurred following foot pronating during heel strike. Some studies have documented a reduction in foot pronation, as well as knee rotation and lateral bend (Eng and Pierrynowski, 1994; Marinakis and Catalfamo, 2004) during the stance phase of gait with insole use. Nester et al. (2003) and Marinakis and Catalfamo (2004), similarly observed a change in foot pronation, but found no subsequent change in knee, hip and pelvis kinematics. This is also supported by the work of Reischl and colleagues (1999), who found no relationship between the magnitude of peak pronation, and peak internal rotation of the tibia or femur. Standing research however has found a relation between foot and pelvis position. Betsch et al. (2011) found that changes in foot position while standing can cause significant alterations in pelvic position, with no difference being found in spine position. This however does not appear to be conclusive, as Duval et al. (2010) found a relationship between internal rotation of the leg and anterior pelvic tilt, but with no link between foot pronation and lumbar, or pelvic moment. The results of Betsch et al. (2011) also goes against the results of Day et al. (1984), and Khamis and Yizhar (2006) that found a direct relationship between pelvic and lumbar position during standing. Similarly, Nelson-Wong and Callaghan (2010) found that sloping inclined and declined surface not only affected pelvic and lumbar angles, but also L_5/S_1 shear and compression. It remains unclear whether insoles can result in a change in spine flexion while walking.

Few studies investigating the effects of insoles on the kinematic chain examine the effects of insoles on additional gait-parameters. While studies such as Eng and Pierrynowski (1994) and Marinakis and Catalfamo (2004) documented changes in lower limb movement following insole use, these studies do not comment on whether the insoles had the same effect on both the left and right limb. Significant differences between left and right limb muscle activity and kinematics can be considered asymmetry (Gundersen et al, 1989), and is of clinical importance assessing gait efficiency (Patterson et al., 2008 and 2010; Burnett et al., 2011). Differences found between left and right limbs following insole use, not present initially, could be interpreted as a negative result of insole use.

In order to determine the benefits of reducing spine flexion and adopting a more neutral, lordotic, lumbar spine posture, the principles of spine loading must be examined. As the external load experienced by the spine is comprised of the weight of the body, as well as the flexion-bending moment, by reducing the amount of spine/trunk flexion there is a reduction in spine loading (White and Panjabi, 1990). Any external load applied to the spine is counterbalanced by ligament and back muscle forces (McGill, 2007). Deformation of the intervertebral disc is a result of both the forces applied by muscles and ligaments, as well as static and inertial body segments (McGill and Norman, 1986). It is the combination of the weight of the body, as well as the ligament and back muscle force that cause compression on the spine. Chaffin (1969) describes the total compressive load applied to the disc as the primary measure of stress to the low back. McGill (2007), expanded on this concept, explaining how the spine is designed for compression in a neutral posture and how it is compression in a non-neutral posture that is harmful to the spine. White and Panjabi (1990), also discussed how this compression is greater during dynamic exercises versus static behaviour, continuing to illustrate the importance of reducing the flexing bending moment during walking. In addition to the compressive stress experienced by the disc and nucleus during flexion/bending, the annular fibres are exposed to tensile stress during the flexion and bending (White and Panjabi, 1990; McGill, 2007). Marras (1993) found a sustained non-neutral flexion angle as a risk factor for developing LBP. While axial twist has little effect on compression, it also increases the tensile stress placed on the annular fibres of the disc and reduces the loading required to cause a disc injury (Drake et al., 2005 and 2008; Tampier et al., 2007). The flexion-bending moment is not the only kinematic measure for predicting LBP during walking. However, by reducing the flexion-bending moment during walking, it is potentially optimizing the load distribution amongst the spine tissues (muscles, ligaments, disc, vertebrae) as well as theoretically optimizing the spine tissues positioning to resist any applied loading (both compression and shear forces) (McGill, 2007), therefore minimizing the risk of injury to the low back.

2.4.4 Footwear and EMG

Due to the lack of consistent kinematic evidence supporting the kinematic chain for reducing LBP, studies have begun to investigate whether these believed changes could have an impact on muscle activation. Studies such as Tomaro and Burdett (1993), Ogon et al. (2001), Bird et al. (2003), and Murley and Bird (2006), displayed altered electromyography (EMG) activity following a footwear intervention. These studies, while exhibiting changes in EMG following an intervention, lack consistency in terms of the shift be a reduction or an increase. Where other studies, such as Sacco et al. (2012), have found no change in muscle activation following similar interventions. Footwear intended to cause strength training during walking has been shown to increase muscle activity of the lower limbs, and therefore has been deemed a useful training method (Romkes et al., 2006). Nigg et al. (2006) also indicated the value of training footwear for its ability to improve ankle and knee strength and proprioception. However, when describing an optimal everyday insole, shoe, or orthotic, Nigg (2001) describes how the product should act to reduce muscle activity. If the insoles act to support the optimal movement path, muscle activity should be reduced. Therefore if an insole causes an increase in activation it is supporting an inefficient path (Nigg, 2001). This emphasizes the importance of distinguishing whether an insole is designed as a training tool, or for everyday use.

Examined EMG for the trunk, as well as lower limbs, is important when analyzing muscle activation during walking. While muscles of the trunk are not required for locomotion, their activation is paramount in maintaining spine stability. During experiments where the spine is isolated from its musculature, it has been documented buckling under a load of 2 kg (Lucas and Bresler, 1961). An everyday insole that increases muscle activity could be viewed as beneficial or detrimental, as the increase could be a result of greater spine instability or could result in greater spine stability. While seeming beneficial, a decrease in muscle activity may be a result of the person supporting themselves more on passive tissues, which is a known injury mechanism (Lühring et al., 2015). Nigg (2006) similarly supports this concept of an increase in muscle activity having the potential to be either beneficial or harmful. An increase in muscle activity may be beneficial if the intention is to temporarily strength train the muscle (Nigg et al., 2006), where if the insole is encouraging the most optimal movement path, it should act to decrease muscle activation (Nigg 2001). Changes in muscle activity may be a result of a change in kinematics, as Heckathorne and Childress (1981) have displayed changes in muscle activation following changes in muscle length. Studies such as Eltoukhy et al. (2012) have been able to relate changes in lower limb muscle activation to changes in foot position. Relatively symmetrical bilateral activation of back muscles is also required during gait in order to slow the movement of the entire torso falling forward following the deceleration of the pelvis during stance (Vink and Karssemeijer, 1988). In addition to the task of locomotion, muscles are also responsible for stabilizing the joints involved in the weight-bearing process of walking (Winter, 1980).

2.4.5 Previous Insole Research Summary

It is unclear whether insole use can act to alter the biomechanics of walking. While questionnaire data has indicated that insole use can significantly reduce LBP, there is little biomechanical research to support this. Insole use may reduce the amount of internal tibia rotation, however they appear to have little impact on knee, hip and pelvis kinematics (Nester et al., 2003; Marinakis and Catalfamo, 2004). In terms of altering muscle activation, the evidence appears to be inconclusive. Research by Tomaro and Burdett (1993), Ogon et al. (2001), Bird et al. (2003), and Murley and Bird (2006), displayed no consistent increase or decrease in EMG following interventions. It is unclear whether insole use decreases LBP due to biomechanical reasons, or if there is some other effect and/or merely a placebo effect.

CHAPTER 3

Study Introduction

3. Introduction

Low back pain is one of the leading causes of disability worldwide, and among the most costly (Lis et al., 2007), with between 60-85% of individuals experiencing LBP at least once in their lives (Bird and Payne, 1999). Due to the lack of understanding regarding the pathology of LBP, several treatments and protocols have been prescribed with little evidence indicating if, or to what extent, they are successful. One example of such treatments are insoles and orthoses (Rothbart et al., 1995), which have been prescribed in large numbers, yet there is little understanding of the mechanism by which these products are to work (Collier, 2011). Hard insoles provide structural support (Cambron et al., 2011; Lockard, 1988), with it being suggested that they can control for abnormal foot motion (Nigg et al., 1999). Soft insoles on the other hand tend to act more as shock absorbers (Ball and Afheldt, 2001; Lockard, 1988; Ogon et al., 2001). The insoles produced by Barefoot Science Canada Ltd. ®, are an adjustable rigid insole, designed for arch strengthening and neuromuscular training (Barefoot Science Products and Services Inc.). Insoles are generally lumped into the above two main categories, but there are several different subtypes of insoles with their own specifically designed benefit, as explained in section 2.4. While many studies have documented a reduction in the reporting of LBP via questionnaires (Dananberg and Guiliano, 1999; Ferrari, 2007; Cambrom et al, 2011), few studies have quantified their effect on EMG and motion analysis, for the lower limbs and spine together.

The research regarding the ability of insoles to reduce LBP appears to be inconclusive. Studies based on questionnaire data have observed a decrease in LBP following insoles use, however there is little quantitative biomechanics evidence for their success. Insoles are commonly designed to reduce the internal rotation of the foot during pronation (Lafortune et al., 1994). This is often in attempt to prevent unnecessary movement from passing up the kinematic chain, to the knee, hip, pelvis and back (Bird and Payne, 1999). Internal rotation of the femur is believed to push posteriorly on the pelvis, causing anterior rotation of the pelvis and lumbar spine (Bird and Payne, 1999; Duvail et al., 2010). Increasing spinal flexion could result in an increase in spine loading, as it would increase the flexion-bending moment (White and Panjabi, 1990). An increase in spinal flexion could also indicate less stability, as the spine should be in neutral position to allow for supporting muscles to be at their optimal length (McGill, 2007) Research by Eng and Pierrynowski, (1994), Marinakis and Catalfamo (2004) and Nester et al., (2003) have displayed decreases in pronation of the foot and internal rotation of the tibia with insoles use. Some of this same research has reported the effects of insoles on knee, hip and pelvis kinematics to be minimal (Marinakis and Catalfamo, 2004; Nester et al., 2003). While Marinakis and Catalfamo (2004) did not observe a reduction in spine flexion, they did however find a reduction in trunk lateral bend with insole use. Even though it is not the primary focus of this study, it is important to determine if these changes act negatively on other gait parameters important to determining efficiency, such as symmetry (Patterson et al., 2008; Burnett et al., 2011). It is still unclear whether the kinematic chain can alter pelvis and low back movement during gait.

The kinematic chain has somewhat been displayed in standing research. Research by Betsch et al. (2011) and Pinto et al. (2008) found changes in pelvic tilt as a result of altered foot position. Similarly, Duval et al. (2010) found a relationship between internal rotation of the leg and anterior pelvic tilt. Furthermore, Day et al. (1984) and Khamis and Yizhar (2006) found a direct relationship between pelvic and lumbar position during standing. This however was not supported by the results of Betsch et al. (2011) and Duvail et al. (2010). There is no conclusive kinematic evidence to support that changes in foot position can alter low back position, as previous research does not typically investigate the kinematic chain starting at the foot, and how it progresses up to the upper back.

Surface EMG has also been used to test the impact of insoles on the biomechanics of the spine. Research by Tomaro and Burdett (1993), Ogon et al. (2001), Bird et al. (2003), and Murley and Bird (2006) displayed altered muscle activity following a footwear intervention. What these studies highlight, as there was no consistent increase or decrease in EMG, that it is unclear whether insoles should increase or decrease muscle activity. The direction of the change appears to rely on the intended purpose of the insoles. Insoles designed for strength training causing an increase in muscle activity may be beneficial, as it may result in an increase in ankle and knee strength, as well as proprioception (Nigg et al., 2006; Romkes et al., 2006). However, Nigg (2001) describes how an insole intended for everyday use should act to reduce muscle activity by encouraging the optimal, or most efficient. Nigg (2001) describes this optimal path as a joints minimal resistance movement. If an insole causes a long term increase in activation during gait, it is supporting an inefficient path (Nigg, 2001). An additional problem with increasing muscle activity is that it can be associated with an increase in joint load (White and Panjabi, 1990). Likewise, the purpose of the insole plays a role in determining if a change in muscle activity and/or kinematics should be considered beneficial or harmful.

The completed study aims to quantify the effects of a particular insole on three-dimensional whole body motion and muscle activation in key muscles in the back, abdomen, pelvis and legs during walking. The purpose of this study was to examine whether a neuromuscular training insole that uses a progression of inserts altered young healthy adults' 3D joint kinematics, and muscle activity after two months of insole use as per the manufacturers guidelines (appropriate progression of inserts). Additional gait parameters may be beneficial in assessing the impact of an insoles on walking biomechanics, however, this study focused on analyzing the effect of the insoles on variables related to the kinematic chain, such as angular range of motion and the mean, minimum and maximum angle observed.

CHAPTER 4

Methodology

4.0 Methodology

4.1 Participants

The participants consisted of 16 subjects (8 males and 8 females) from a university-aged population. Male participants on average (±*SEM*) were 25 years old (2.88), 1.82 m (0.02) tall and had a mass of 83.21 kg (1.77), and females on average were 24.6 years old (4.75), had a height of 1.65 m (0.03) and a mass of 63.23 kg(3.19). Participants were excluded based on the following selection criteria: currently using prescribed orthotics, any 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 work or school; any previous back, hip, leg, or foot surgery; inability to stand for more than four hours; and an inability to walk for more than 60 minutes. Informed consent was obtained for each participant prior to data collection, and York University's Office of Research Ethics approved all protocols.

4.2 Instrumentation/Equipment

Participants were asked to wear the provided insoles during their everyday life for eight weeks and to follow the manufacturer's instructions regarding the progression through the levels of inserts. By everyday life, participants were instructed to wear the insoles at all times during every activity that did not require a specific type of footwear that was not compliant with the insoles. The insoles consisted of the Barefoot Sciences® Products and Services Inc. (Mississauga, Canada), ¾ length, active model, and were provided to each participant based on shoe size. The Barefoot Science® insoles are soft, but have spacing for inserts that vary in density and so stiffness, that allow users to follow a progression from the softest/smallest/shortest insert to the hardest/longest/tallest insert (Figure 4.1). Participants were instructed to follow the manufacturer's guidelines. In brief, the manufacturer's information instructed users to begin at level one and take one week to progress from one insert level to the next, if they felt any discomfort after a full day's use they were instructed to return to their previous insole. Participants who progressed to level 4 or beyond were classified as the compliant group (C). Participants who progressed to level 3 or less were classified as non-compliant (NC). This division was made as it was believed the participants who proceeded past the half way point in levels would display additional benefits with insole use compared to the participants who did not make it past level 3. It was also used to determine if participants who reported to wear the product more, or used the product during more physically exerting tasked, reached a higher insert level.

Figure 4.1. Barefoot Sciences®, ¾ length, active model insoles. Inserts are organized left to right, progressing from the softest/smallest/shortest insert (level 2), to the hardest/longest/tallest insert (level 6). The level 1 insert was placed inside the arch support (blue circle) of the upside down insole at the top of figure.

Muscle activation was recorded from 12 muscles bilaterally, using three AMT-8 EMG amplifier systems (Bortec Biomedical Ltd., Calgary, Canada) (Figure 4.1). Surface EMG was collected from the following muscles: thoracic erector spinae, 2.5 cm lateral to T_9 spinous process; lumbar erector spinae, 3 cm lateral to L_3 spinous process; rectus abdominis, 3 cm lateral to the umbilicus; external obliques, 15 cm lateral to umbilicus; internal obliques, superior to inguinal ligament; and latissimus dorsi, lateral to T_9 spinous process (Drake et al., 2006). Additionally gluteus medius, midway between the greater trochanter and the sacrum (Nelson-Wong and Callaghan, 2008); gastrocnemius medialis, half way between the medial side of the popliteus cavity to the medial side of the Achilles tendon insertion (Rainoldi et al., 2004), tibialis anterior, between lower margin of the patella and lateral ankle (Zipp, 1982); peroneus longus, a quarter of the way between the tip of the head of the fibula and the tip of the lateral malleolus (Hermens et al., 2000), biceps femoris, between ischial tuberosity and the lateral epicondyle of the tibia (Hermens et al., 2000); and vastus medialis, 80% of the way between the anterior superior spine of pelvis and medial gap of the knee (Hermens et al., 2000). Three reference electrodes, one for each AMT-8 system, were placed on the left and right clavicle and left knee. Each participant's EMG data were normalized to their maximal voluntary contraction level (%MVC) (see section 4.4). EMG signals were differentially amplified (frequency response 10-1000 Hz, common mode rejection 115 dB at 60 Hz, input impedance 1000 G-Ω; model AMT-8, Bortec, Calgary, Canada) and converted at 2400 Hz (Vicon MX, Vicon Systems Ltd., Oxford, UK) from analog to digital form.

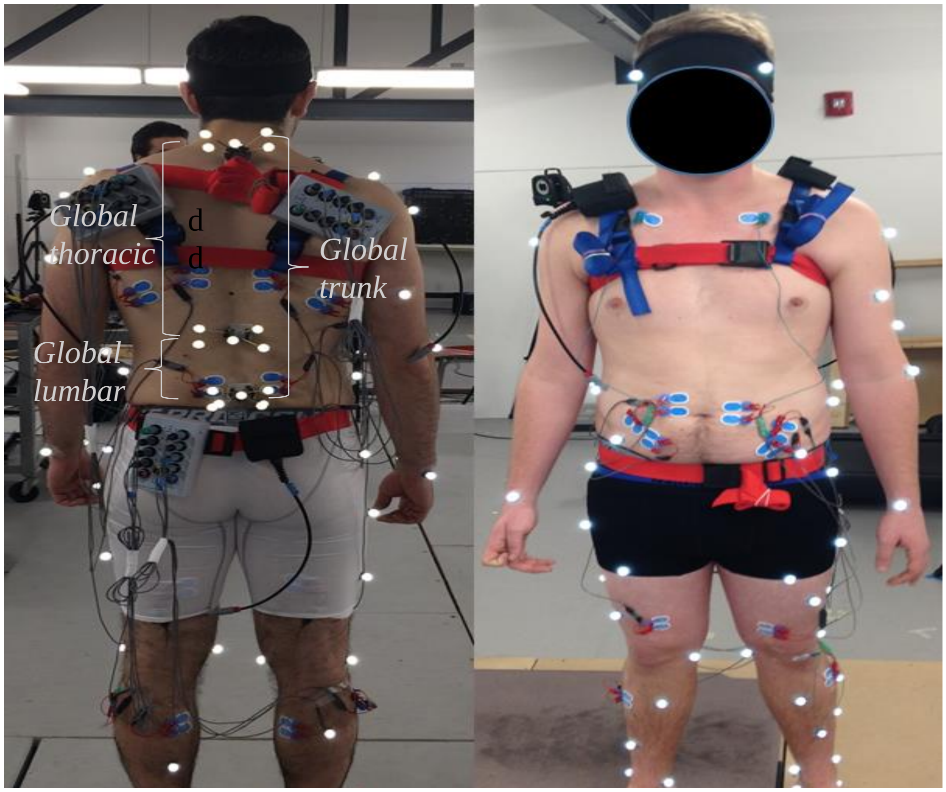

Figure 4.2. Represents the posterior and anterior view of marker and electrode placement for the back and abdomen. With back marker clusters being placed on the T_1, T_{12} and L_5 vertebra (top, middle and bottom), making up the *Global thoracic, lumbar and trunk segments.*

Kinematics were recorded at 50 Hz, using a seven-camera motion capture system (Vicon MX, Vicon Systems Ltd., Oxford, UK). Seventy-six reflective markers were adhered to the skin using double sided tape, in a full-body model configuration as shown in Figure 4.2 with the locations detailed in Table 4.1. A walkway 8 m long and 1 m wide was constructed in the centre of the motion capture space. All walking trials were performed walking from one end of the walkway to the other.

Throughout the eight week period, a weekly questionnaire was used to track participants' advancement through the insoles progression. Questionnaires were sent out to participants at the same time via weekly email. The questionnaire monitored the participant's weekly insole level, insole use, physical activity, average time spent sitting, walking and standing, as well as physical well-being (Figure 4.3). The questionnaire also included questions to indicate the participant's self-assessment for physical well-being. Participants were asked to circle on a scale of 1-10, how they felt physically (1=worst, 10=best) (Figure 4.3). For the purpose of this study, participants' insert level progression, as well as hours of daily use, were the primary focus.

Table 4.1. Passive reflective marker locations by body region. All marker locations are bilateral with the exception of the head and trunk.

Body region	Passive Reflective Marker Location*
Head	1 Right Front Head* 1 Left Front Head* 1 Right Back Head* 1 Left Back Head* 1 Middle Back Head*
Shoulder	1 Acromion* 1 Shoulder (Centre of Rotation)*
Upper Arm	2 Tracking Markers
Elbow	1 Humerus Medial Epicondyle* 1 Humerus Lateral Epicondyle*
Forearm	1 Tracking Marker*
Wrist	1 Ulna Styloid Process* 1 Radius Styloid Process*
Trunk	1 C_7* 3 Spine clusters (5 markers each) T_1 (Dunk and Callaghan, 2005) T_{12} (Dunk and Callaghan, 2005) L_5 (Dunk and Callaghan, 2005)
Pelvis	1 Iliac Crest* 1 ASIS: Anterior Superior Iliac Spine* 1 PSIS: Posterior Superior Iliac Spine*
Thigh	4 Tracking Markers* 1 Femur Greater Trochanter*
Knee	1 Femur Medial Epicondyle* 1 Femur Lateral Epicondyle*
Shank	1 Achilles Tendon* 2 Tracking Markers*
Ankle	1 Tibia Medial Malleolus* 1 Fibula Lateral Malleolus*
Foot	1 Second Meta-tarsal (Leardini et al., 2007) 1 Heel* 1Lateral Foot*

* indicates that marker placements were taken from Livingston (2007)

Weekly Questionnaire

Participant Code: ______________________________ Date: ________________________

1. What level insert do you currently have in your insole? (circle one)
   1 2 3 4 5 6 7

2. Did you increase the insert level over the past seven days? (circle one) Y N

3. On average during the past seven days, how long did you wear your insoles each day?
   ____

4. Typically, what activity/activities did you participate in while wearing the insoles? (circle all that apply)
   Sitting
   Standing
   Walking
   Activities of daily living (for example, household chores)
   Weight training
   Cardio training
   Other

5. On average, how much time did you spend at each activity per day over the past seven days?
   Sitting ____________________________
   Standing ____________________________
   Walking ____________________________
   Activities of daily living (for example, household chores) ____________________________
   Weight training ____________________________
   Cardio training ____________________________
   Other ____________________________

6. On a scale of 1-10, how do you feel physically? (1 = worst, 10 = best) (circle one)
   1 2 3 4 5 6 7 8 9 10

Comments:
____
____
____
____

Figure 4.3: Copy of the questionnaire used to track the weekly progression of the participant through the insert program. The questionnaire will also be used to track participant`s physical activity level throughout the week.

4.3 Procedures

Participants completed two in-lab data collection sessions, eight weeks apart. The first collection served as the Initial-pre and the second collection served as the Final-post and was completed following eight weeks of insole use. Muscle activation and 3D whole body motion was collected for all walking trials.

Prior to electrode placement, data collection was started by obtaining the circumference and anthropometric data required for the Vicon Nexus Plug-in Gait analysis (Vicon Systems Ltd., Oxford, UK): weight (kg), height (m), bilateral leg lengths (cm), bilateral knee widths (cm), bilateral ankle widths (cm), and trunk depth. Trunk depth, knee width and ankle width were collected using calipers, leg lengths were recorded using measuring tape. Skin preparation for electrode placement was completed according to the typical EMG collection protocols (McGill, 1991; Mirka and Marras, 1993; Drake et al., 2006; Nelson-Wong and Callaghan, 2010), including shaving of hair and alcohol swabbing over electrode placement sites. Electrodes were adhered according to section 4.3, with locations near the centre of the muscle bellies to allow for the best observable signal (Gilmore and Meyers, 1983). Following electrode placement, the participant completed a 5 minute quiet-rest trial, with the participant resting supine. The collected EMG represented the baseline activation for each muscle.

Following the rest trial eleven different tasks were completed to collect the MVC for the muscles being observed in the study. Table 4.2 describes the isometric contractions each participant was required to hold for three seconds. A participant completed two trials of each MVC, with over a minute break between trials. MVCs were used to obtain each participant's maximum muscle activation for normalization of EMG (See Data Processing).

Table 4.2: Represents each muscle being observed for EMG, along with their MVC trials and its reference

Muscle	MVC	Reference
Thoracic erector spinae	Lying on the bench, back extension	McGill, 1992
Lumbar erector spinae	Lying on the bench, back extension	McGill, 1992
Rectus abdominis	Sitting on the bench, crunch	McGill, 1992
External obliques	Sitting on the bench, twist	McGill, 1992
Internal obliques	Sitting on the bench, bend	McGill, 1992
Latissimus dorsi	Pull down	Arlotta et al., 2011
Gluteus medius	Resisted hip abduction in the side lying position	Nelson-Wong et al., 2008
Tibialis anterior	Dorsiflex about the ankle against manual resistance through a range of motion while lying supine.	Schinkel-Ivy et al., 2012
Peroneus longus	Resisted Pronation	Murley et al., 2009
Biceps femoris	Prone resisted knee flexion, knee flexed at 55°	Rutherford et al., 2011
Vastus medialis	Restricted prone knee extension, knee already at 15°	Rutherford et al., 2011
Medial gastrocnemius	Knee extended plantarflexion	Murley et al., 2009

Following MVCs, the participant was equipped with reflective markers in order to capture 3D-kinematics. Participants then completed an upright standing trial, where participants were asked to stand in the centre of the capture space and were instructed to stand as upright as possible, feet shoulder width apart, imagining a string pulling the top of their head towards the ceiling. The upright standing trial was completed to allow for the participants upright posture to be subtracted from their posture during all walking trials.

Participants were then asked to complete 10 walkway trials. Trials consisted of participants walking across the instrumented walkway, with synchronous collection of surface EMG and 3D-kinematics. Figure 4.4 displays a mock-up of an instrumented participant walking across the walkway in the centre of the capture space. Walkway trials were repeated if participant`s consecutive foot falls did not land in the centre of their respective plate. Participants were allowed practice walks prior to collecting in order to get used to the length of the walkway and to ensure no instrumentation was encumbering. All walkway trial data files were trimmed to the three strides performed in the centre of the walkway in attempt to not include initiation and termination, as well as to insure stable accuracy of the capture volume.

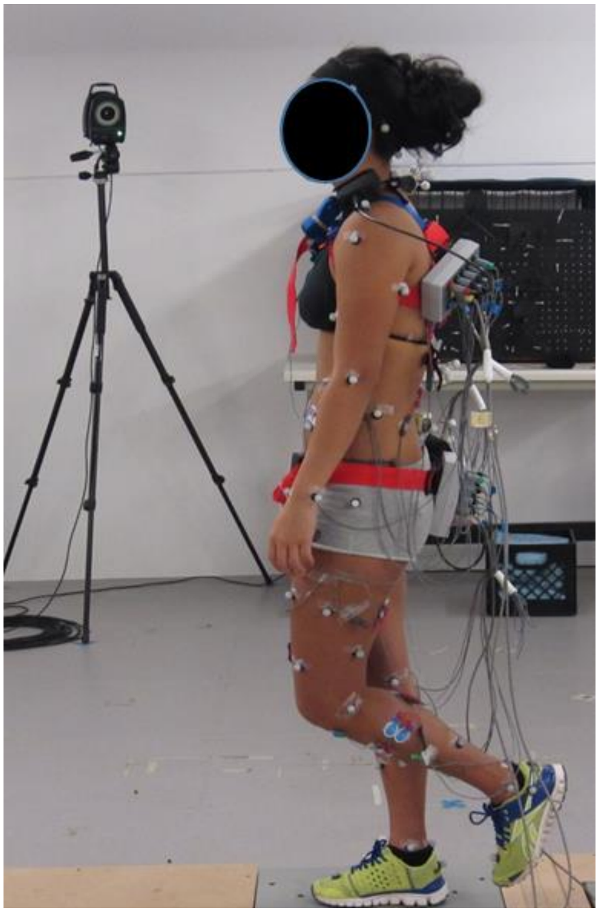

Figure 4.4. Represents a participant walking across the capture space while being monitored for surface EMG and 3D Kinematics

4.4 Data Processing

All data processing for kinematics and kinetics was done using Visual3D (C-Motion Inc., Germantown, MD). The EMG data was initially high pass filtered at 30 Hz to remove electrocardiogram contamination (Drake and Callaghan, 2006). These data were then full wave rectified and dual-pass filtered with a fourth-order low-pass Butterworth filter using a cut-off frequency of 6 Hz (Winter, 2009), producing a linear envelope for each of the 24 muscles being recorded. From the 5 min rest trial, a 30 s window representative of rest was selected based on the linear envelope profile and used to calculate each muscles resting mean activation, which was then subtracted from all subsequent signals. The linear envelope profile was windowed in attempt to minimize signal spikes or artifact across all of the muscles. The signals were then normalized to a percent of maximum muscle activation, using the participant's MVC trials. Burden et al. (2003) stated that normalizing the muscle activation data in a gait cycle to %MVC is the best method of normalization that allows for the interpretation of to what degree a muscle is active. Using the normalized data, the mean activation was quantified for each muscle.

Kinematic data were dual-pass filtered using a fourth-order, low-pass Butterworth filter with a cut-off frequency of 6 Hz (Winter, 2009). Segments were created and were called the *global thoracic spine*, *lumbar spine* and *trunk*, as well as *pelvis*, *thigh*, *shank*, and *foot.* These segments were defined as global regional angle, which is the segment's angle relative to the lab spaces. Each global segment's upright position from the initial standing trial was subtracted out to represent upright as 0°. All three-dimensional rotation angles were calculated using an X-Y-Z (Flexion-Lateral Bend-Axial Twist) Cardan sequence (Preuss and Popovic, 2010). Based on these global segments, relative joint angles were created for the *thoracic spine* (thoracolumbar junction, T_{12}-L_1), *lumbar spine* (Lumbosacral junction, L_5-S_1), *trunk*, *hip*, *knee* and *ankle*.

Angles for each joint were expressed as relative and global angles, with a segment's relative angle being defined as that segment relative to the segment below (Table 4.3). Angles for left and right limb stance phase were multiplied by -1 when appropriate so that the directions and references would represent the same movement for both limbs stance phase (Table 4.3). For each trial, the ROM and mean angle during each legs' stance phase were compared pre- and post-intervention. Considering that a change in both the maximum and minimum angle observed in the same direction would not result in a change of ROM, the maximum and minimum angles were also compared pre- and post-intervention. This study focused on stance phase as maximum pronation has been reported to occur within the first 50% of stance (Nawoacsenski and Ludewig, 1999). As described in section 2.2.1, insoles often attempt to reduce the amount of trunk flexion via a reduction in foot pronation, located several links down the kinematic chain. Further, approximately 80% of the gait cycle is also comprised of single stance phase (Winter, 2009), indicating its importance. Therefore, the ROM of each segment was defined as the difference between the absolute largest angles observed during the trial and the absolute minimum angle observed during the stance phase.

Table 4.3: Summary of all global and relative segments, including each axis and respective movement. Twisting towards stance limb indicates axial twist, with the anterior aspect of the segment rotating so it faces the stance

Angle	Primary Segment	Relative to	X Motion	Y motion	Z Motion
Global Segments (Foot, Shank, Thigh, Pelvis, Trunk, Thoracic, Lumbar)	Specific Segment	Lab	(+)Anterior tilt (-)Posterior tilt	(+)Tilt towards Stance Limb (-) Tilt towards Swing Limb	(+)Twist towards Stance Limb (-) Twist towards Swing Limb
Ankle	Shank	Foot	(+)Dorsi flexion (-)Plantar flexion	(+)Eversion (-)Inversion	(+)External Rotation (-)Internal Rotation
Knee	Thigh	Shank	(+)Extension (-)Flexion	(+)Valgus (-)Varus	(+)External Rotation (-)Internal Rotation
Hip	Pelvis	Thigh	(+)Flexion (-)Extension	(+)Abduction (-)Adduction	(+)Internal Rotation (-)External Rotation
Trunk	Trunk	Pelvis	(+)Flexion (-)Extension	(+)Side bend towards stance limb (-)Side bend towards swing limb	(+)Twist towards Stance Limb (-) Twist towards Swing Limb
Lumbar	L5	Pelvis	(+)Flexion (-)Extension	(+)Side bend towards stance limb (-)Side bend towards swing limb	(+)Twist towards Stance Limb (-) Twist towards Swing Limb
Thoracic	T1	L1	(+)Flexion -Extension	(+)Side bend towards stance limb (-)Side bend towards swing limb	(+)Twist towards Stance Limb (-)Twist towards Swing Limb

4.5 Data Analysis

A four-way mixed multivariate analysis of variance (MANOVA) was used to analyze both the observed mean EMG and mean angle together, for all walking trials. The two repeated factors used in the analysis were insole condition (pre, post) and side (left, right limb stance phase) and the two between group factors were sex and compliance. With muscle activation being known to change with muscle length (Heckathorne and Childress, 1981; Eltoukhy et al. 2012) it is expected that a change in muscle activity would result in a change joint angle and vice versa. Each muscle was paired with its primary respective joint based on MVCs, with a separate MANOVA being run for each kinematic axis. The mean EMG for the muscles on the stance limb side of the body were calculated for each limbs stance phase separately, in order to compare stance of the left limb to stance of the right limb. Similarly, the mean angle observed represented the average angle for each segment on the stance limb side of the body, with a separate average being calculated for each right and left stance phase. The means were an average of the three strides observed for each trial, averaged across all trials. All significant MANOVA results were further analyzed using a similar four-way repeated measures analysis of variance (ANOVA) detailed below, examining each variable separately (Clemente et al., 2014). All analyses were considered significant at α=0.05, with significant omnibus *F*-tests being further analyzed pairwise, and the *p*-values being adjusted using a Bonferroni correction. For all statistical analysis, only interactions or main effects involving the factor of visit were of particular interest to this study. Other significant interactions would not be of use in determining the effectiveness of the insoles in altering EMG or kinematics. A biostatistician with the York

University Statistical Consulting Service was consulted to insure that this MANOVA procedure would be an acceptable analysis for both EMG and kinematics.

All segments kinematics were analyzed using an ANOVA as well, regardless of whether MANOVA results indicated a change in both joint kinematics and muscle activation, to determine if the insoles trial period only altered the kinematic measures. For all maximum, minimum, ROM and mean angles the same four-way mixed ANOVA was applied, with the X, Y and Z axes being separated, with a separate mean calculated for each segment of interest. The two repeated factors used in the analysis were insole condition (pre, post) and side (left, right limb stance phase) and the two between group factors were sex and compliance. For all walking trials, a maximum, minimum, ROM and mean angle was calculated by averaging each variable across all three strides of the walk, with the final mean averaging all trials from a given condition. For all statistical analysis, only main effects of visit, or interactions involving a factor of visit, were of interest in displaying differences pre-/post-trial period.

Similar ANOVAs were also run on all respective muscles, regardless of MANOVA results, to determine if the insoles had an effect on just muscle activation and not kinematics. The normalized stance limb mean EMG for each muscle, from all walking trials were analyzed using a four-way mixed ANOVA. Once again, the two repeated factors used in the analysis were insole condition (pre, post) and side (left, right limb stance phase) and the two between group factors were sex and compliance. In addition to this, a supplementary five-way mixed ANOVA, separate from the MANOVA, was run on torso muscles, with the additional factor looking at both the stance and swing side of the body. The additional factor of side was to determine if muscles on both sides of the torso were activated equally during stance, as suggested by Vink and Karssemeijer (1988) in section 2.4.4

CHAPTER 5

Results

Results

Appendix section contains pre-/post-trial graphs for all segments and muscles results, displaying the ROM, mean, maximum and minimum angle, for kinematic data, and mean %MVC for EMG data. These graphs do not represent the results of the statistical analyses performed as they display the results by visit, regardless of interactions or main effects being present. Only graphs of statistical significance are presented in the results section. For a full statistical summary of all kinematic and EMG interactions and main effects please see Appendix B, as only interactions of interest, interactions containing a factor of visit, are examined in this results section.

5.1 Population Characteristics

Table 5.1 displays the mean (±*SEM*) insert level reached and daily use for participants following the eight week trial period.

Table 5.1. Summary of the average (±SEM) daily use in hours (h/day), and the final insert level (L), after the eight week trial. Means are displayed for the entire population, non-compliant group and compliant group, as well as for each group broken down by sex. n=represents the sample size of each group.

	Entire Population	Non-compliant	Compliant
Male	L 3.50 (0.60) for 5.66 h/day (0.75), n=8	L 2.25 (0.43) for 5.90 h/day (1.32), n=5	L 5.33 (1.53) for 5.40 h/day (3.06), n=3
Female	L 3.63 (0.53) for 8.47 h/day (1.36), n=8	L 2.50 (0.50) for 7.29 h/day (1.95), n= 4	L 5.00 (0.50) for 10.03 h/day (1.93), n=4
Total	L 3.56 (0.39) for 7.06 h/day (0.83), n=16	L 2.38 (0.31) for 6.60 h/day (1.09), n=9	L 5.17 (0.28) for 7.68 h/day (1.35), n=7

According to questionnaire data, participants reported the majority of their insole use during tasks requiring walking and sitting, with eight participants using their insoles during their normal weight training, in addition to activities of daily living (ADLs) and 11 participants using their insoles during their normal cardiovascular activities, in addition to ADLs. Those who progressed to a level 4 insert and above were classified as compliant, where those that did not where classified as non-compliant. A total of seven individuals (4 female, 3 male) progressed past level three, with the female compliant group wearing the product the most. Nine individuals were classified as non-compliant (4 female, 5 male). The female complaint group wore the product on average ≈2.7 h/day longer than the non-compliant group. Males in the non-compliant group on the other hand, wore their insoles for approximately the same amount of time daily as the compliant group, differing by ≈0.5 h/day. Daily use appeared to have no indication on whether participants used their insoles during more physically demanding tasks. As eight members (4 females, 4 male) of the non-compliant group used the insoles during their regular cardiovascular workouts and five participants (3 female, 2 male) used the insoles during their regular weight training. Whereas for the compliant group, only three participants (1 female, 2 male) wore them during their normal cardiovascular workout, and three participants (1 females, 2 male) wore the insoles during weight training. Considering participants generally reached level three or four over the 8 week period, participants typically took 2 weeks to progress from one insert level to the next.

5.2 MANOVA

Examining the results of the MANOVA, that compared means for both EMG and angle observed pre-/post-trial together, interactions of interest were only found for lumbar spine angle/activation. With a separate MANOVA being run for each plane, a side*visit*sex*level interaction was detected for lumbar erector spinae mean EMG, with frontal ($F(1,23)=3.725, p=0.040$) and sagittal plane lumbar spine motion ($F(1,23)=3.903, p=0.035$). All MANOVA interactions were investigated further using ANOVA results for EMG and kinematics separately. This was done to determine if the interaction was a result of a change in both EMG and kinematics, or just a large enough change in one variable to detect an interaction. ANOVA results for lumbar erector spinae activation displayed a sex*side*level*visit interaction ($F(1,12)=7.196, p=0.020$), with the non-compliant female left limb stance phase increasing from 2.33%MVC (0.34) to 5.48%MVC (1.68) ($p=.016$)(Figure 5.1). Comparing this to kinematics, no similar ANOVA interactions, or any interactions involving visit, were detected for mean *lumbar spine* motion in either plane ($F(1,15)<2.222, p>0.157$). Examining all other segments/muscle MANOVA results, no interactions or main effects of interest were detected ($F(1,23)<3.139, p<0.062$). The insoles appeared to have no significant effect when it came to altering EMG and joint angle together.

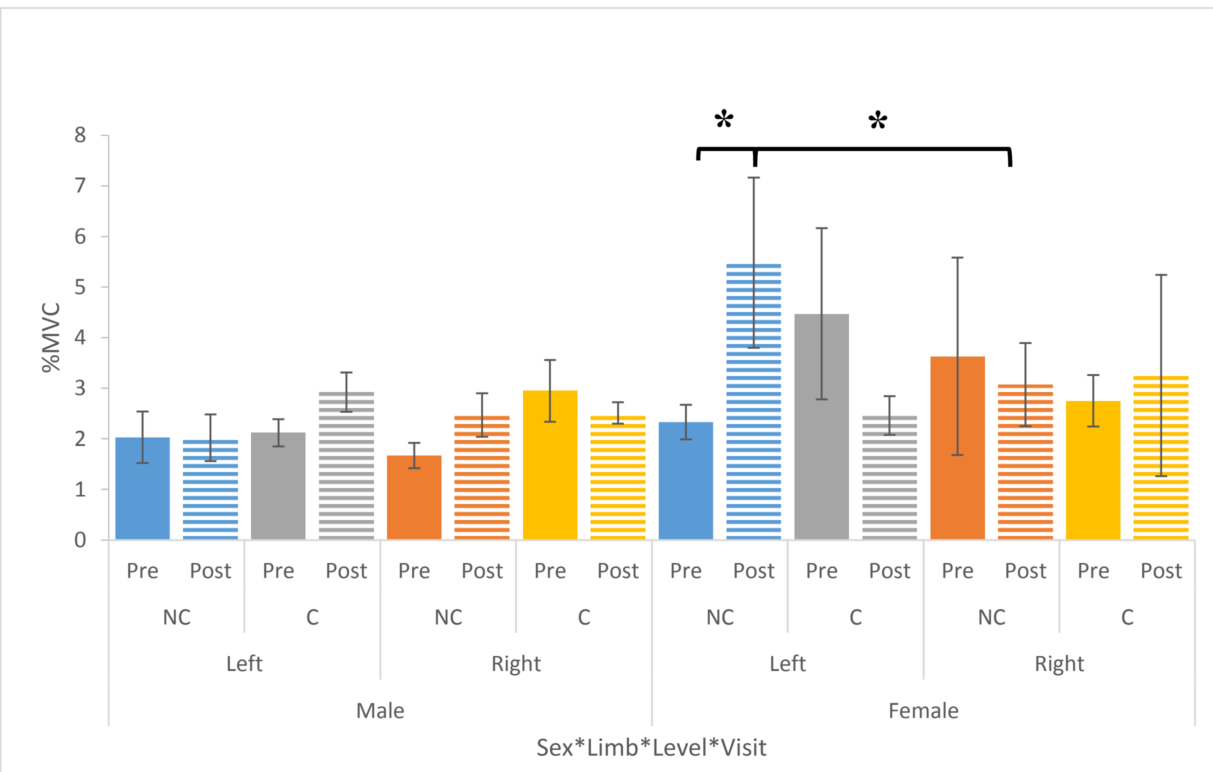

Figure 5.1. Lumbar erector spinae activation expressed to display the sex*limb*level*visit interaction observed in the ANOVA analysis. The solid fill represents the initial visit, with the pattern fill representing the post visit. Left foot non-compliance, or NC (blue), and compliance, or C (grey) groups, were separated from right foot non-compliance (orange) X and compliance (yellow) groups due to the interaction detected. * Indicates a significance of <0.05.

5.3 Kinematics

The main kinematic outcomes of interest for this study were the ROM, maximum and minimum angle, and the mean angle, that were obtained during both left and right limbs' stance phase for all three planes. Segments that displayed a main effect or interaction involving visit were also of particular interest for this study. Interactions involving compliance and visit are discussed in this section separately from interactions involving sex and visit, or just a main effect of visit. Interactions involving compliance and visit were separated since they are not generalizable to the entire population, as the average person would be unaware if they belong in the compliant or non-compliant group. In addition to these results, all interactions involving visit that did not result in a post-hoc displaying differences pre-/post-trial are stated at the end of this section.

The spine was the only area of the body that exhibited changes in the ROM that occurred during stance phase pre-/post-trial. Except for frontal plane motion in *thoracic* and *global trunk*, where both segments displayed an interaction between sex and visit ($F(1,14)>6.91, p<0.020$) (Figure 5.2), no main effect, or interaction of interest were found for all other segments frontal plane ROM ($F(1,15)<2.638, p>0.125$), ($F(1,14)<2.896, p>0.111$), ($F(1,13)<2.478, p>0.139$) Examining the sex*visit interaction for frontal plane *thoracic* ROM, males exhibited an increase of 2.47° ($p=0.045$), while females displayed a decrease in *thoracic* lateral bend by 4.01° ($p=0.003$) (Figure 5.2). Where for the global trunk segment only females displayed a difference pre-/post-trial, showing a decrease in ROM by 0.39° ($p=0.02$). The insoles appeared to have little effect on joint ROM following the trial period.

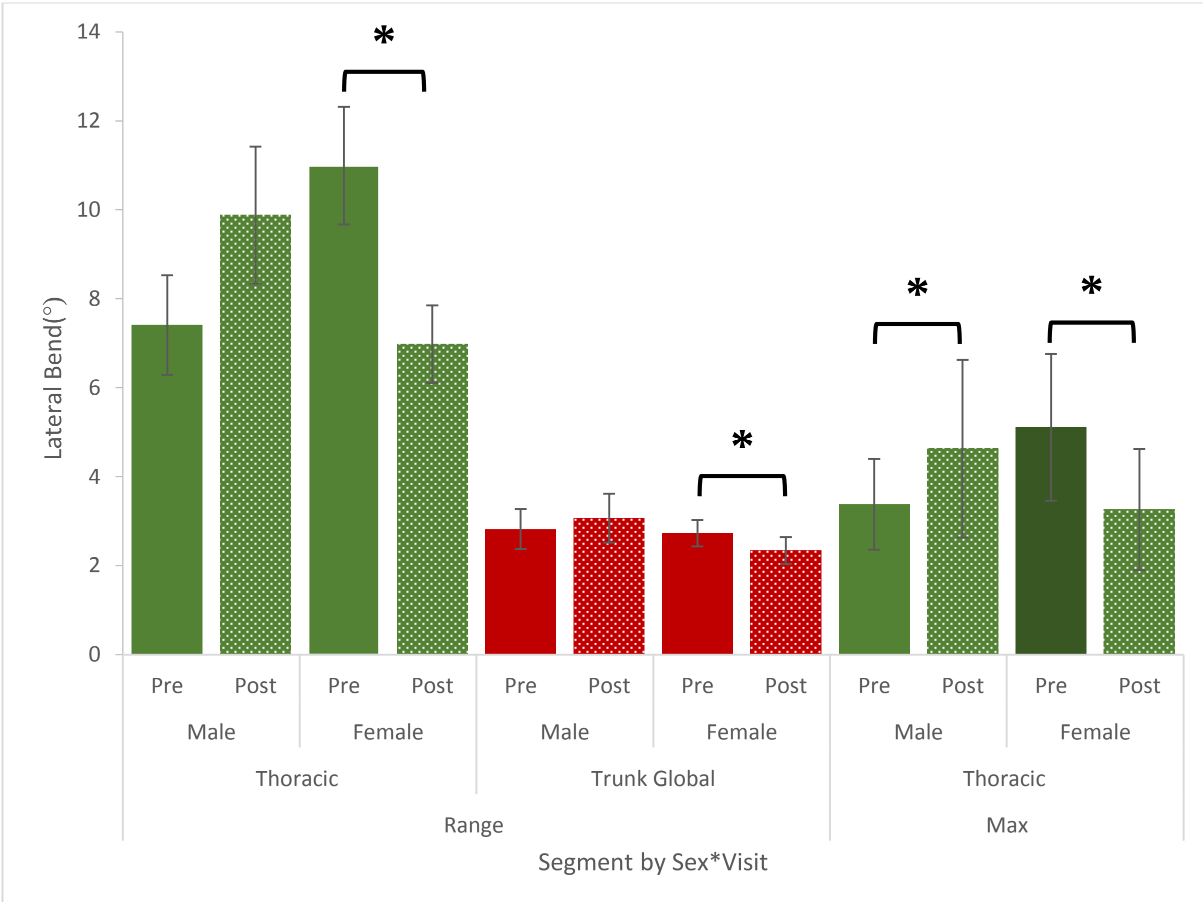

Figure 5.2. Frontal plane motion for thoracic maximum and range, as well as trunk range of motion. Displays the sex*visit interaction for frontal plane Thoracic (green) range and maximum, as well as Trunk (red) maximum. (+) values indicate lateral bend towards the stance limb, where (-) values indicates lateral bend towards swinging limb. The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

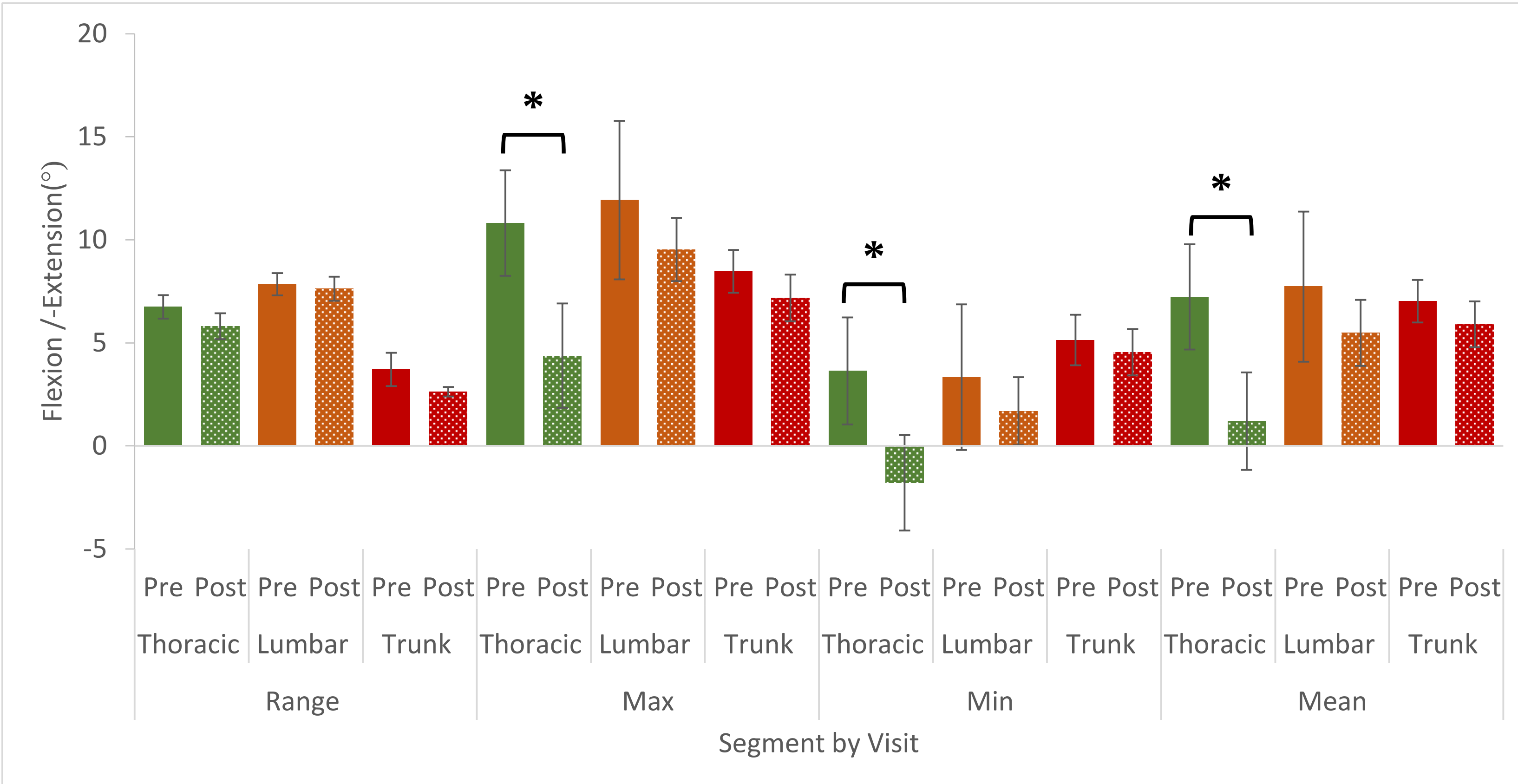

Figure 5.3. Sagittal plane range of motion, as well as maximum, minimum and mean angle observed for Thoracic (green), Lumbar (brown) and Trunk (red) motion. The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

Considering that a similar shift in both the maximum and minimum angle observed would not result in a change of overall ROM, the maximum and minimum angles were also compared. For the example of sagittal plane spine movement, positive numbers represented flexion and negative values represented extension, indicating that the maximum would be the greatest amount of flexion, and the minimum would be the greatest amount of extension, The only significant difference found for both the maximum and minimum angle was for the *thoracic spine* in the sagittal plane (Figure 5.3). The maximum angle observed a difference in pre-/post-trial for thoracic flexion (p=0.004), which was reduced by approximately 60%, from 10.81° (2.56) to 4.38° (2.53). Similarly, the minimum angle observed also displayed a difference during stance (p=0.019), changing from 3.64° (2.59) flexion to 1.79° (2.31) extension (-1.79° flexion). Mean sagittal plane *thoracic* motion also showed a difference post-trial (p=0.004), reducing the mean flexion to approximately 17% of its original value, from 7.23 (2.55) to 1.20 (2.37). Although the differences were not significant, lumbar and trunk segments also appeared to have a reduction in mean, maximum, and minimum angle. Only *thoracic* flexion exhibited significant changes pre-/post-trials for mean, maximum, and minimum angle observed, although additional significant differences were found for spine segments when examining mean, maximum or minimum on their own. Along with the changes experienced in the spine stated above, general differences were detected for certain spine segments maximum and mean motion. In the frontal plane, a sex*visit interaction was detected for maximum *thoracic spine* angle (F(1,14)=15.500,p=0.001)(Figure 5.2), as well as a side*visit interaction for mean *trunk* angle (F(1,15)=5.188,p=0.038). Other than the differences in *thoracic spine* and *trunk* movement listed above, no additional interactions or main effects of interest were detected for the general population's global and relative *trunk*, *lumbar*, *or thoracic spine* maximum, minimum or mean angle, in any direction ($F(1,15)<3.595, p>0.077$) ($F(1,14)<4.233, p>0.059$) ($F(1,12)<1.586, p>0.233$). Frontal plane *thoracic spine* motion displayed a sex*visit interaction, with males displaying a difference pre-post-trial ($p=0.005$), increasing from 3.38° (1.02) pre-trial, to 4.63° (2.00) post -trial. Females conversely displayed a difference ($p=0.042$) as a result of a decrease in maximum frontal plane *thoracic spine* angle, decreasing from 5.11° (1.65) to 3.26° (1.36) post-trial. Frontal plane mean trunk angle displayed a change for the right legs stance phase ($p=0.009$), from 0.805° (0.35) right lateral bend, to 0.83° (0.28) left lateral bend. Therefore, in addition to the changes experienced in the sagittal plane, the insoles appeared to affect frontal plane motion of the *thoracic spine* and *trunk* regions.

Differences were detected for *knee* and *shank* kinematics, following the trial period. A side*visit interaction was detected for frontal plane maximum *shank* angle ($F(1,15)=6.075, p=0.026$), as well mean *knee* angle ($F(1,14)=6.119, p=0.027$) (Figure 5.4). Sex*visit interactions were detected for mean ($F(1,13)=4.762, p=0.048$) and minimum ($F(1,13)=7.406, p=0.017$) sagittal plane *knee* angle (Figure 5.5). No additional interactions, or main effects of interest were found for lower limb mean, minimum and maximum angle observed ($F(1,15)<2.893, p>0.110$) ($F(1,14)<3.990, p>0.066$) ($F(1,13)<1.827, p>0.199$). Female participants' minimum ($p=0.036$) and mean ($p=0.014$) sagittal knee angle changed following the eight week trial period, decreasing from -47.14° (2.05) to -45.25° (1.41) and from -13.93° (1.92) to -11.19° (0.95). The side*visit interactions for the shank and knee were detected in the frontal plane. Examining the left legs stance phase, a significant difference was detected pre-/post-trial for the mean knee angle ($p=0.017$), changing from 1.38° (0.46) valgus to 1.14° (0.74) varus. Frontal shank motion exhibited a difference for maximum lateral bend during right limb stance phase following the trial period ($p=0.014$), shifting from 5.02° (0.45) to 6.49° (0.49). No other differences were detected for the entire sample's kinematic ROM, mean, maximum and minimum angle observed following the trial period. The trial period appeared to contribute to a reduction in female knee flexion and a change in frontal plane knee and shank motion.

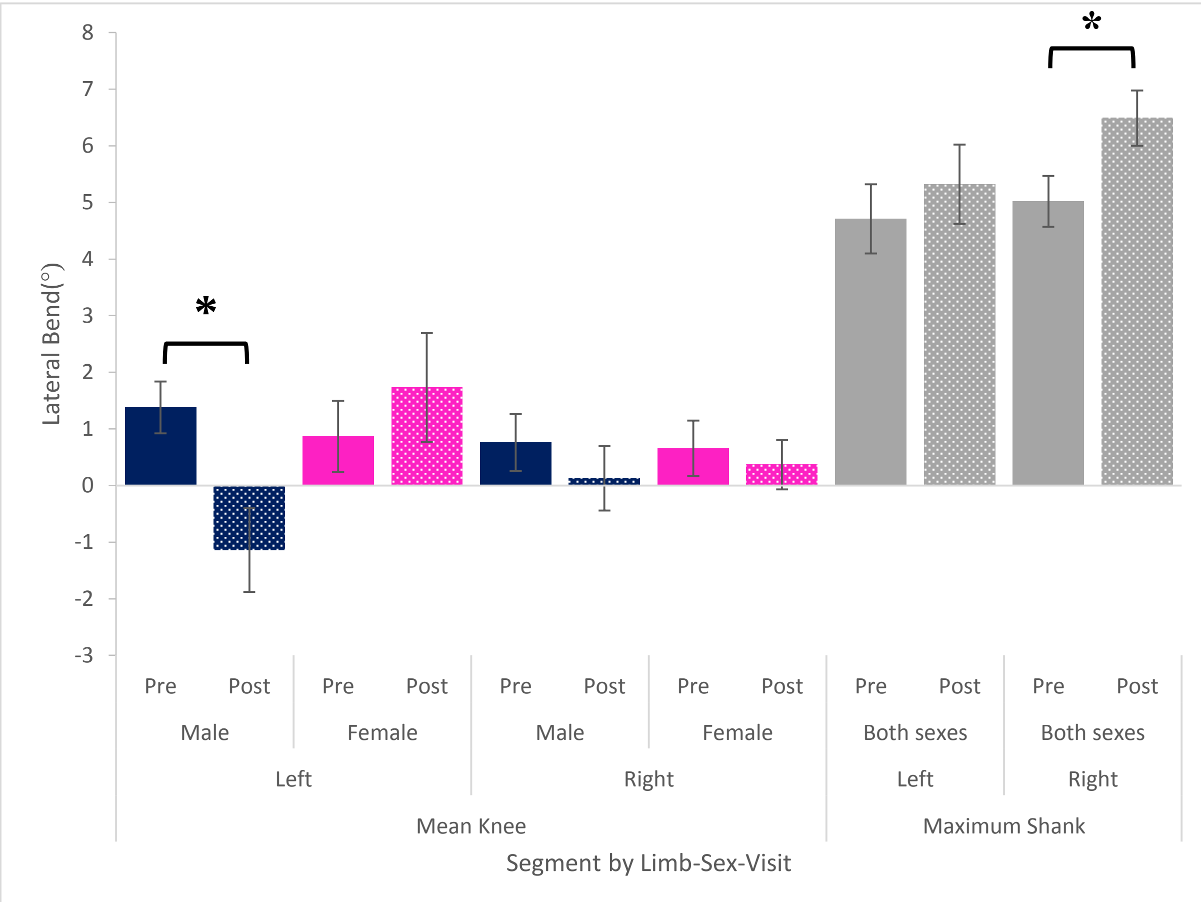

Figure 5.4. Additional frontal plane interaction with Males (blue) and Females (pink) mean knee displaying a limb*sex*visit interaction and maximum shank displaying a limb*visit interaction collapsed across sex (grey). (+) values indicate lateral bend towards the stance limb, where (-) values indicates lateral bend towards swinging limb. The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

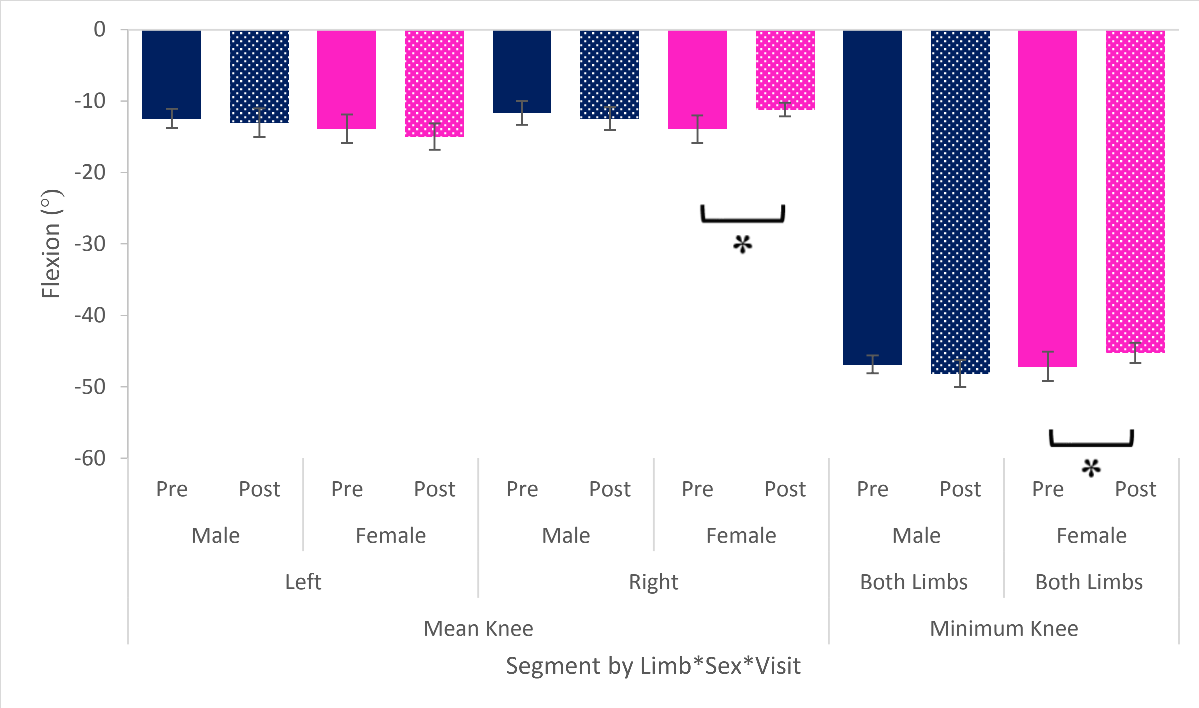

Figure 5.5. Additional sagittal plane limb*sex*visit interactions for mean and minimum knee angles, with males (blue) being separate then females (pink). The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

Multiple kinematic variables exhibited interactions involving compliance and visit, with post-hoc tests indicating differences between pre and post collections. Both the hip and pelvis, displayed compliance*visit interactions for the mean and minimum angle observed (F(1,14)>4.745,p<0.047)(Figure 5.6). The non-compliant group showed a decrease in the minimum hip flexion angle, from -23.92° (1.55) to -26.65° (1.39)(p=0.023), and an increase in minimum pelvic tilt from -3.75° (0.91) to -2.59°(1.03)(p=0.030). Means for the non-compliant group showed similar changes. Mean hip flexion changed from -3.15° (1.48) to -5.30° (0.71)(p=0.044), with the mean for the pelvic tilt increasing from -2.19° (0.96) to -0.43° (0.79)(p=0.038). Side*compliance*visit interactions were detected for maximum transverse plane trunk angle (F(1,13)=6.356,p=0.026) and mean sagittal plane shank angle (F(1,13)=5.529,p=0.035) (Figure 5.7). The non-compliant group exhibited a change in maximum transverse plane angle during left limb stance phase (p=0.042), decreasing from 3.15°

(0.63) to 1.77° (0.31). Conversely, the shank displayed changes for mean sagittal plane angle during right limb stance phase (p=0.047), decreasing from 9.23° (0.89) to 7.87° (0.94). All differences between visits were detected for the non-compliant group only.

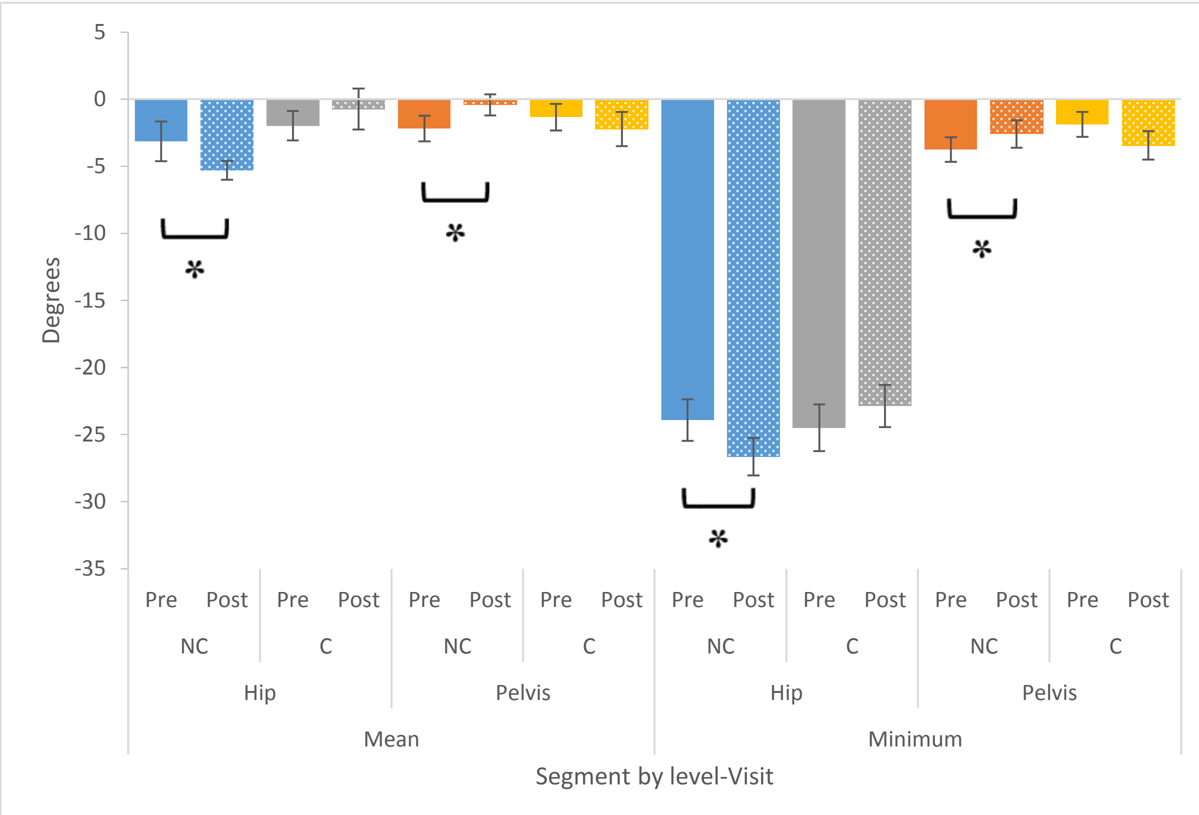

Figure 5.6. Level*visit interactions for mean and minimum angle observed for sagittal plane pelvis (blue for non-compliant and grey for compliance) and hip motion (Orange for non-compliance and gold compliance). The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

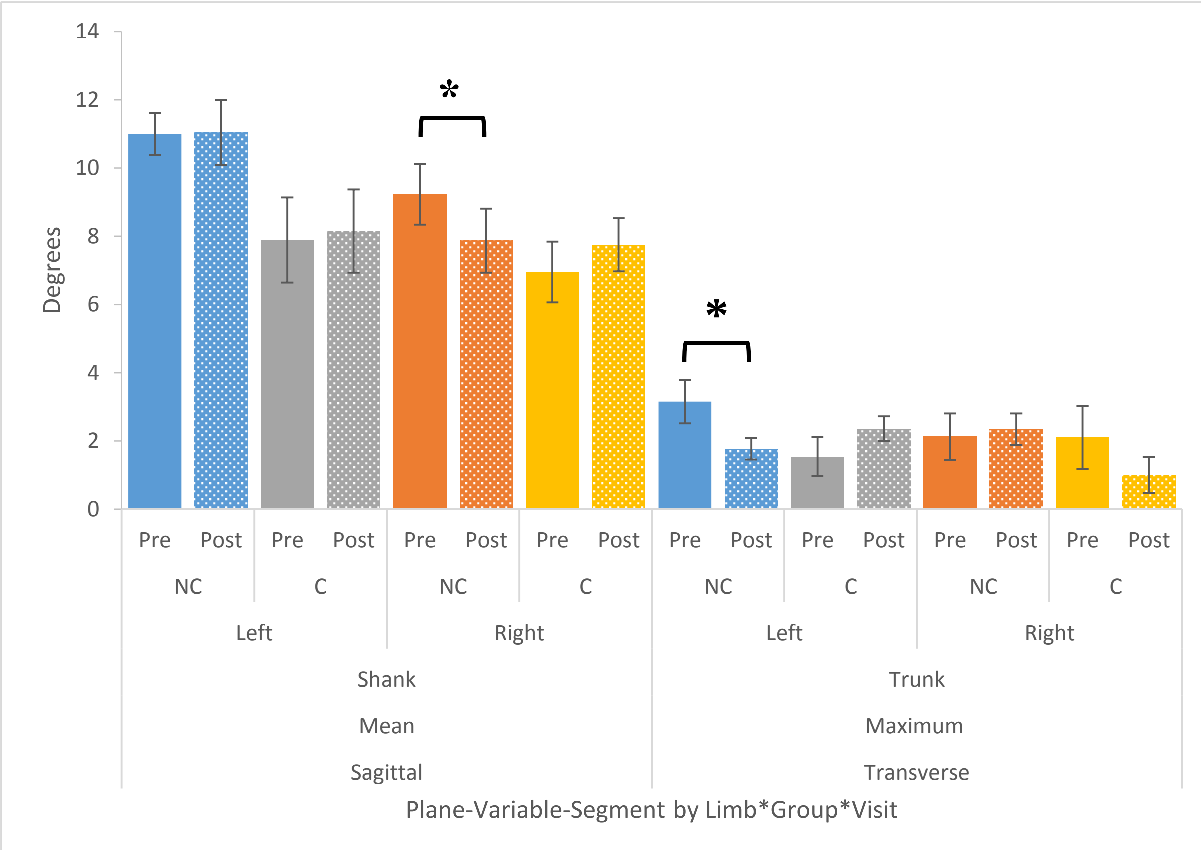

Figure 5.7. Displays the limb-group visit interaction for mean sagittal shank angle and maximum transverse trunk angle separated into Left non-compliant (blue) and compliant (grey), as well as right non-compliant (orange), and compliant (gold). With * indicating a significance of <0.05.

Multiple interactions involving visit were detected for the mean and maximum angle observed that were not of interest to answering the primary research question. There were multiple examples of side*visit interactions that only resulted in significant differences between sides for one visit, as well as two interactions that resulted in no significant post-hoc results. A side*visit interaction was detected for mean sagittal knee (F(1,13)=4.762,p=0.048), pelvis (F(1,14)=9.723,p=0.008) and global trunk angle (F(1,12)=4.984,p=0.045), as well as maximum global trunk flexion (F(1,13)=7.940,p=0.015) and shank lateral bend angle (F(1,14)=15.500,p=0.001). For all of the side*visit interactions a significance difference was detected between participants post-trial left and right limbs' stance phase (p<.012) (Figure 5.8-

9). In terms of the interactions that resulted in no significant post hoc results (p>0.069), a side*visit interaction was detected for maximum global lumbar sagittal plane tilt (F(1,15)=7.160,p=0.017), mean hip flexion (F(1,14)=4.475,p=0.047) and a side*sex*visit interaction was displayed for mean *thoracic* twist (F(1,14)=3.612,p=0.033). In summary, all of the side*visit interactions with significant post-hoc results indicating differences between limbs post-trial were not present following the initial collection.

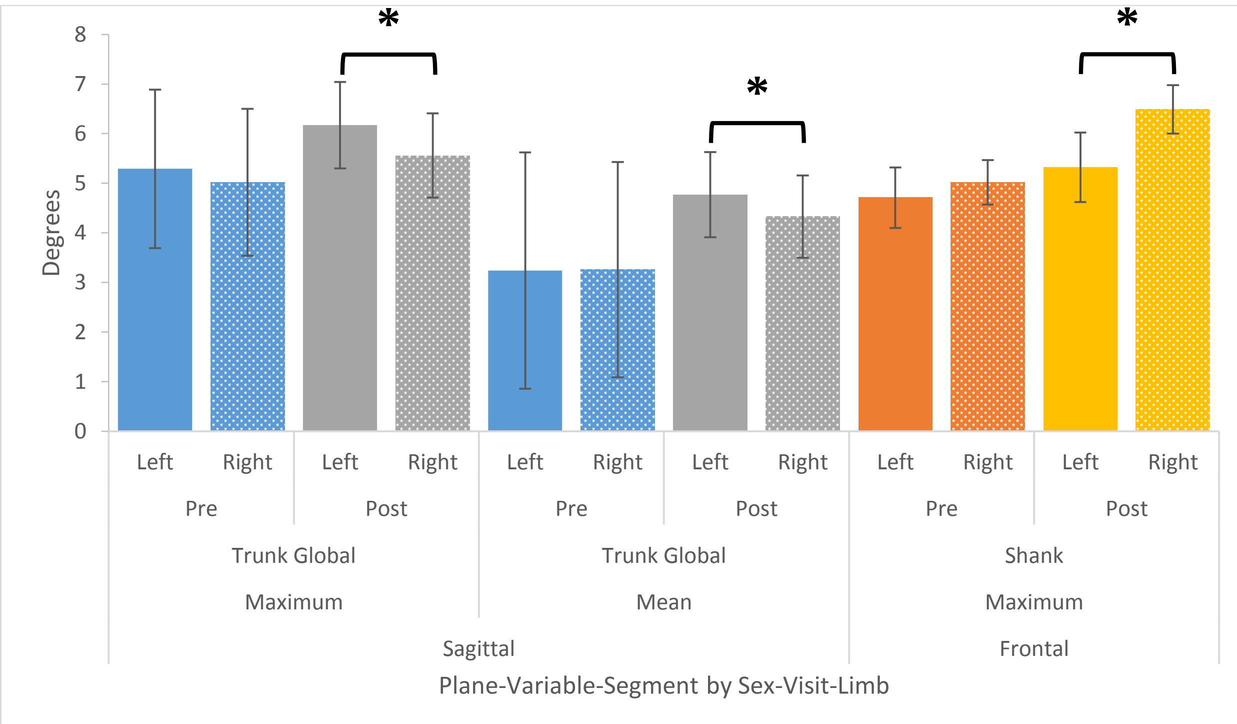

Figure 5.8. Sagittal plane mean and maximum global trunk angle, as well as frontal plane shank maximum, collapsed across sex. Illustrates the side*visit*limb interaction for the Trunk Pre (blue) and post (grey) visit, as well as the shanks pre (orange) and post (gold) visit. Solid fill indicates the value for the left stance phase, where a pattern fill represents the right stance phase. For the global trunk left and right just indicates whether the angle was observed during right stance phase or left stance phase. The * indicating a significance of <0.05.

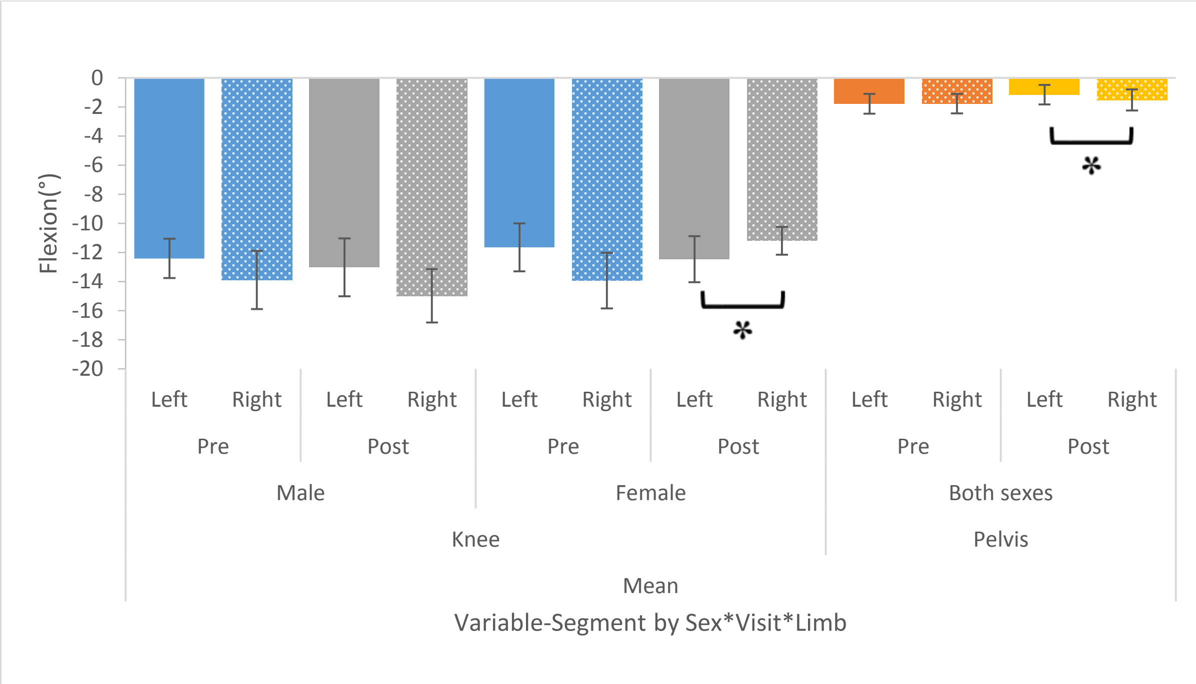

Figure 5.9. Sagittal plane mean knee and pelvis motion. Displays the sex*visit*limb and visit*limb interactions for the for knee pre (blue) and post (grey) visit, as well as the pelvis pre (orange) and post (gold) visit. Knee motion was further subdivided by sex with the solid fill indicating the left limb mean, and pattern fills representing the right limb mean. With * indicating a significance of <0.05.

5.4 EMG

The analyses of the EMG data detected very few differences pre-/post-trial for mean %MVC during the stance phase of walking. Gastrocnemius, lumbar erector spinae, tibialis anterior and vastus medialis, all displayed a main effect or an interaction effect of visit. Significant differences between visits were identified with post-hoc tests for the vastus medialis and lumbar erector spinae. Vastus medialis displayed a main effect of visit ($F(1,15)=5.411, p=0.034$), with EMG increasing from 5.17%MVC (0.82) to 6.16%MVC (0.97) ($p=0.034$) (Figure 5.10). As stated in the MANOVA section (5.1), lumbar erector spinae displayed a side*visit*sex*level interaction ($F(1,12)=7.196, p=0.020$), with non-compliant female left limb stance phase increasing from 2.33%MVC (0.34) to 5.48%MVC (1.68) ($p=0.016$) post-trial(Table 5.1). A difference was also detected between the muscles left and right stance phase post-trial ($p=0.040$) that was not observed during the initial pre collection

(p=0.318). Examining the side*visit interaction exhibited by gastrocnemius ($F(1,14)=6.044, p=0.028$), a difference of ≈6%MVC was found between participants pre-trial left and right limbs stance phase (p=0.022), that was reduced to ≈1 %MVC following the trial period (p=0.069) (Figure 5.11). While tibialis anterior resulted in a level*visit interaction ($F(1,12)=5.642, p=0.035$), post-hoc tests resulted in no difference ($p>0.074$). With exception of the four muscles listed above, no other interactions or main effects were found involving visit ($F(1,15)<3.495, p>0.081$) ($F(1,14)<1.256, p>0.281$). Vastus medialis activation was the only difference in mean EMG found across all participants.

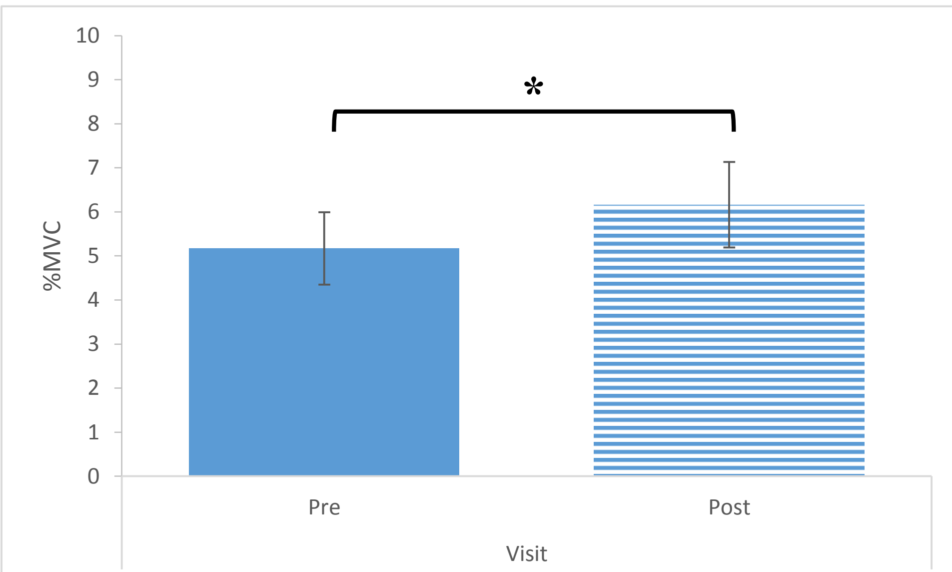

Figure 5.10. Displays the vastus medialis mean EMG (%MVC) for the pre (solid) and post (pattern) visit. With * indicating a significance of <0.05.

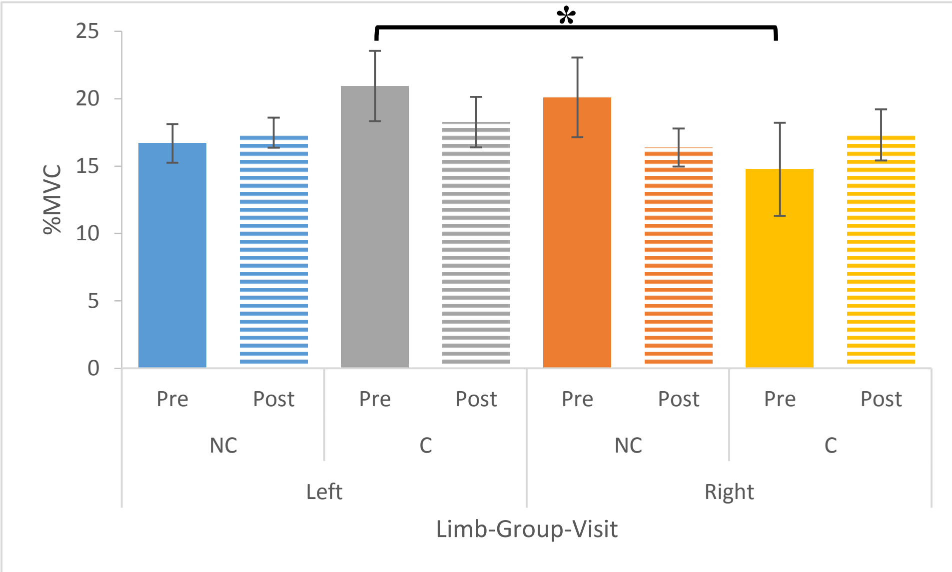

Figure 5.11. Displays Gastrocnemius mean EMG (%MVC) for the left limb non-compliant (blue) and compliant (grey) group, and the right limb non-compliant (orange) and compliant (gold) group. The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

The supplementary five-way mixed ANOVA, which factored in the torso muscles on the swinging limb's side of the body, had little difference to the original ANOVA. Similarly, the lumbar erector spinae displayed a side*visit*sex*level interaction ($F(1,12)=7.205, p=0.020$), with non-compliant females left limbs stance phase increasing from 2.54%MVC (0.39) to 5.76%MVC (1.64) $p=0.012$). A step*visit*sex interaction was also detected for the external oblique muscles (Figure 5.12). After insoles use females displayed a difference between stance and swing side, external oblique activation. The difference between the sides increased from ≈0.64% pre-intervention ($p=0.511$), to ≈2.45% MVC post-intervention ($p<.001$). No addition interactions, or main effects, involving visit were detected $F(1,15)<2.568, p>0.130$), ($F(1,14)<3.170, p>0.097$). Examining both the left and right torso muscles instead of just the stance limbs side of the body had no impact on the interpretation of the results compared to the initial analysis.

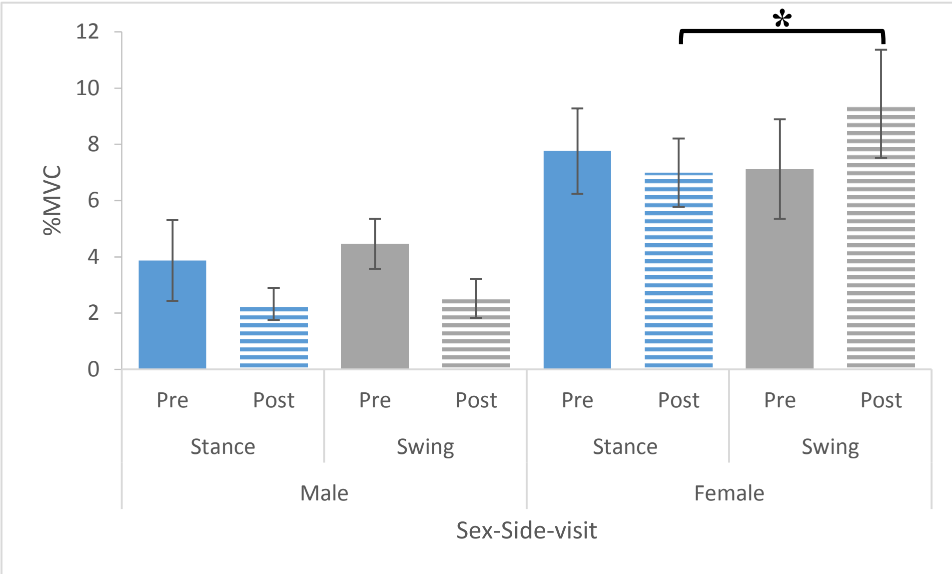

Figure 5.12. Displays the sex*side*visit interaction for the External Oblique mean surface EMG (%MVC), for stance (blue) and swing (grey) phase. The solid fill represents the initial visit and the pattern fill representing the post visit, with * indicating a significance of <0.05.

CHAPTER 6

Discussion

Discussion

This study was the first study to quantify the long term effects of insole use on muscle activity and 3D kinematics for the lower limbs and torso simultaneously. Also this was the first study to separate the general trunk/spine angle into *lumbar* and *thoracic* spine segments for a gait related analysis. Kinematic and EMG data appeared to be within a normal range expected during gait for an unaffected university aged population. ROM for the *ankle*, *knee* and *hip* appeared to be similar to those reported by Kadaba et al. (1990) for a similar population. This study displayed baseline sagittal, frontal and transverse plane ROM of ≈20°, 10° and 10°, respectively, for the *ankle*; 45°, 8° and 7°, respectively, for the *knee*; and 37°, 9° and 7°, respectively, for the *hip*. Where Kadaba et al. (1990) found ROM to be ≈25° and 15° for *ankle* sagittal and frontal ROM. For sagittal, frontal and transverse motion, Kadaba et al. (1990) reported ROMs of ≈52°, 12° and 13° for the *knee*, and 40°, 7° and 10° for the *hip*. Saha at al. (2008), reported similar sagittal plane motion as well, with *ankle*, *knee* and *hip* motion to be ≈26°, 63° and 35°. *Pelvis* and *lumbar spine* ROM reported by Thurston and Harris (1983), was reported to be ≈4°, 11° and 15° for *pelvis*, and 5°, 11°, and 8.3° for *lumbar spine*, sagittal, frontal and transverse, ROM. This study reported pre-trial values of ≈3°, 5° and 8° for the pelvis and 8°, 5°, and 8° for the *lumbar spine* ROM respectively. Similar to the results of Burden et al. (2003), muscle activity during walking was generally lower than 20 %MVC, with no mean activation for the population being greater than 20 %MVC during the study. Likewise, the muscle activity and 3D kinematics collected are considered to be similar to the values reported in the literature.

6.1 Effects on Mean EMG and Kinematics

The insoles did not significantly alter the relationship between the mean activation and angle observed, regardless of joint or axis of motion. Interactions were found for the EMG values paired with frontal plane and sagittal plane kinematics, however neither of these interactions resulted in a significant ANOVA that would indicate changes for both joints muscle activation or angle observed. Considering there were some changes found for either EMG or kinematics, it is possible that over the eight week period that these changes offset each other or masked changes when the data were paired. For example the insoles could have had an immediate effect on the lower thoracic erector spinae muscle activation, contributing to the decrease in thoracic flexion. Considering muscle activation has been known to change with changes in muscle length (Heckathorne and Childress, 1981) and research by Eltoukhy et al. (2012) has documented the immediate change in EMG muscle activity as a result of altered foot positions, it is unlikely there would be a change in kinematics without a corresponding change in EMG, unless the change was too small to be statistically significant. The insoles could have caused an increase in muscle activation initially, perhaps after one or two weeks, but this could have become masked following eight weeks of use due to neuromuscular adaptation. This could have resulted in no change in mean EMG and even potentially a reduction in EMG if the trial period was longer than eight weeks. Improved efficiency of muscle activation or increased strength over the eight week period could result in a lower %MVC while maintaining the same posture. The reduction in female *knee* flexion, as well as the increase in vastus medialis activation, was not large enough to reveal a significant main effect or interaction of visit. If there was a main effect of visit between the two, it could be inferred that the increase in vastus medialis activation contributed to the reduction in *knee* angle exhibited by females post-trial. It is possible that if participants were observed immediately following the start of insole use, after one week of insole use, or potentially every week during insole use, a significant difference would have been detected. However this type of participant commitment is very difficult to attain. Therefore, while it was important to quantify the effects of eight weeks of insole use, it is important that we also quantify the immediate effects of insoles in order to determine their full effect on both kinematics and muscle activation.

6.2 Reduction in Thoracic Spine Kinematics Angles

The thoracic spine's frontal plane movement was the only segment to display significant changes across multiple variables following the trial period. *Thoracic spine* maximum, minimum and mean flexion angle all experienced decreases post-trial. While no difference was detected for *thoracic spine* sagittal plane ROM, this was a result of the maximum and minimum angle changing by a similar magnitude, in the same direction. Mean, maximum and minimum *thoracic* flexion was reduced from their respective initial value by ≈59.5%, 149.2% and 82.4%. There was no significant decrease in the overall *trunk* or *lumbar spine* angle accompanying the decrease in *thoracic* flexion, although values appeared to indicate less flexion. This could be a result of the *trunk* angle also taking into account *pelvis* and *lumbar spine* movement, which only displayed minor, non-significant decreases following the trial period; or the segment being a straight line from the *pelvis* to the shoulders, not necessarily passing through the actual location of the spine. The changes in sagittal plane *thoracic spine* movement, coupled with the *trunk* and *lumbar spine* results, appear to be sufficient evidence to suggest that the participants' spine was in a more upright position following the trial period. Even though the literature has no documented case of reduced spine flexion through insoles use, most examine *trunk* and *lumbar* angle, calculated from the *pelvis* to the shoulders, or *pelvis* to L_1, and so is not sensitive to regional changes. Considering the work of White and Panjabi (1990), and McGill (2007), the adoption of a more upright (less kyphotic thoracic region) could act to reduce the axial load

(Briggs et al., 2007) as well as increase the stability of the thoracic spine and by corollary, or close association, the lumbar spine.

The external load experienced by any functional spine unit within the spine is a result of the compression due to the weight of the body above the spine unit, as well as flexion bending moment of said weight (White and Panjabi, 1990). White and Panjabi (1990) further detail how the flexion bending moment is a result of the weight of the body above the axis, multiplied by the perpendicular distance of the centre of gravity from the axis. Therefore, by reducing the amount of thoracic flexion occurring during walking, the insoles are acting to reduce the perpendicular distance between the centre of gravity and the axis of interest, the thoracolumbar junction. This would also result in a smaller internal spine load, as the flexion bending moment, as well as the anterior shear force, must be counterbalanced by the ligament and back muscle forces. The muscle forces in turn apply compression to the functional spine unit, which further increases stability. While this change is occurring midway up the spine, approximately 40% of an individual's weight is located superior to this location (Winter, 2009). As the thoracic spine connects to the lumbar spine through the T12-L1 functional spine unit, the thoracolumbar junction would experience less loading as a result of reduced thoracic flexion.

Reducing the external load experienced by the thoracolumbar junction would be of significance, as it is a spine level that is particularly vulnerable to injury (White and Panjabi, 1990). Transition points of the spine are particularly susceptible to mechanical failure, as changes in structural properties are stress concentration points (White and Panjabi, 1990). The T_{12}-L_1 functional spine unit is a stress concentration point and according to White and Panjabi (1990), and has been suggested to have one of the highest frequency of spine injuries. Reducing the amount of compression on the spine may help to lower the incidence and reoccurrence rate of these injuries. An insole that reduces the thoracic spine flexion bending moment may also have additional value in preventing and treating low back pain by increasing the stability of the region. McGill (2007) discusses how an upright spine could be an indicator of neuromuscular control by creating equal tension between support muscles, causing the spine to not bend in any one direction.

The changes reported in the literature of lower limb kinematics being associated with a less flexed spine during gait were somewhat present during this study. Research by Saha et al. (2007) indicated that increased *trunk* flexion is also characterized by an increase in *knee* and *hip* flexion during the stance phase of gait. After eight weeks of insole use females exhibited a general decrease in maximum *knee* flexion by ≈2°. Accompanying this change in maximum angle, females exhibited a reduction in mean right limb stance phase *knee* flexion by ≈2.7°. The decrease in *hip* flexion was of a similar magnitude, but was only present for the non-compliant group, with the mean and minimum flexion decreasing by ≈2.7° and ≈2.2°, respectively. The fact that no results, for any group, indicated an increase in *hip* or *knee* flexion post-trial also suggested participants did not exhibit a lower limb movement pattern typical with an increase in spine flexion. Although the changes were small in magnitude, and only found for the female population, they only act to reinforce that there was a decrease in thoracic spine flexion following insole use.

Contrary to the report of Bird et al. (2000) that a reduction in spinal flexion occurs through changes in lower limb kinematics, there were no differences found in lower limb kinematics to suggest the existence of the chain. The expected end result of decreased spinal flexion did occur in the thoracic region, however there was little to no evidence to suggest this was a result of changes in the lower limbs, *pelvis*, or *lumbar spine*. The only evidence to support this idea of the kinematic chain would be the change in male's left limb stance phase. During left limb stance, the mean frontal plane *knee* angle changed from ≈1.4° valgus, to ≈1.1° varus. While there was an increase in the right *shank's* lateral tilt during stance, increasing by ≈1.5°, there was no change in *knee* or *ankle* lateral bend movement. This study is not the first to detect a change in frontal plane *knee* kinematics following an insole trial period (Eng and Pierrynowski, 1994). Bird et al. (2000) suggested that reducing knee valgus should result in reduced spine flexion, through reductions in internal femur rotation and anterior pelvic tilt. This did not appear to be the case in the current study as there was no reduction in *thigh* or *hip* axial twist, as well as *pelvic* anterior tilt. If anything, the opposite occurred, with the non-compliant group showing maximum posterior *pelvic* tilt angle, as well as their mean *pelvis* flexion angle, exhibiting more anterior tilt post-trial compared to the compliant group. A difference was found post-trial with the minimum *pelvis* and mean *pelvis* angle exhibiting ≈0.75° (p=0.023) and ≈1.75° more anterior pelvic tilt, respectively. These data do not act as direct evidence to suggest the inaccuracy of the Bird et al. (2000) kinematic chain theory, as the difference in *knee* angle was found for the male group, while the difference in *pelvis* angle was found for the non-compliant group. In addition, changes of just a few degrees, although statistically significant, may not be functionally relevant for the lower limbs (Eng and Pierrynowski, 1994). These differences may not be functionally relevant for changing lower limb kinematics, however they could have been a factor contributing to the decrease in spine flexion. This notion of insoles having minimal effect on *knee* and *hip* kinematics has been found previously in the literature (Nester et al., 2003; Marinakis and Catalfamo, 2004). In general, changes experienced at one joint did not appear to result in, or be a result of, a significant difference at the next joint in the chain. While this could be evidence against the kinematic chain, these results could also be due to slight changes not manifesting in significant differences, or differences in study populations. Nonetheless, the current thesis study findings do not support the theory of the kinematic chain being the cause for decrease in *thoracic spine* flexion following insole use.

In addition to the differences found in sagittal plane *thoracic spine* kinematics, other differences were found between pre- and post-trial *thoracic spine* frontal plane movements. Frontal plane ROM changed following the trial period, with both sexes displaying post-trial differences in opposite directions. Males exhibited an increase, changing from ≈7.4° to ≈9.9° post-trial period. Females on the other hand, decreased from ≈11° to ≈7°. This could be partly due to a similar significant shift in the maximum frontal plane angle for the *thoracic spine*, with males increasing by ≈1.3° and the females decreasing by ≈1.9°. The larger decrease in female ROM is most likely due to a similar change in the minimum angle that was not found to be statistically different. This reduction in *thoracic* ROM would also explain the decrease in *global trunk* sagittal plane ROM for females, which was smaller due to the *lumbar spine* ROM remaining similar to the values observed in the initial collection. Similar to spine/trunk flexion, side bending is also known to increase the amount of loading, experienced by the spine (White and Panjabi, 1990). The change in mean *trunk* lateral bend during right limb stance phase would have little impact on loading, as the change was from ≈-0.8° to ≈0.8°. Although the insoles appear to be contributing to a reduction in thoracic loading for females, for males the answer does not appear as clear. The decrease in *thoracic* flexion experienced following insoles use was much larger than the increase in *thoracic* lateral bend exhibited by males post-trial, indicating the potential for loading to still be decreased with insole use, for both males and females.

There was also a small change in the maximum angle observed for transverse plane *trunk* movement. The reduction in the non-compliant group's maximum amount of trunk twist towards the stance limb could also be an indication of increased spine stability. Maximum *trunk* twist for the non-compliant group decreased from ≈3.15° to ≈1.77°. It is however impossible to determine if this is a result of the individuals who were unable to progress through the insoles, or the specific level of insole itself. Although reducing the amount of axial twist experienced by the *trunk* may have little impact on compression, it would result in annulus fibres experiencing less shear force (White and Panjabi, 1990). While there is no evidence indicating improvement in lower limb kinematics, the insoles generally tended to decrease the amount of excess movement experienced by the spine during walking.

Differences were present between participants that complied with the insole level progression, versus those that did not. It was expected that participants that progressed further through the insole progression would experience greater changes in walking kinematics than the non-compliant group. Surprisingly, in all cases of a compliance-visit interaction, post-hoc tests only resulted in differences between visits for the non-compliant group. Sagittal plane mean and minimum angle observed for *hip* and *pelvis* motion, as well as mean *shank* motion all experienced changes post-trial in the non-compliant group. Left stance phase *trunk* motion in the transverse plane also experienced changes, with the maximum angle being reduced post-trial. It is possible that the individuals in the compliant group already had the foot type, or optimized neuromuscular control, the insoles were trying to encourage, and therefore would not have received as much of a benefit from insole use.

The examination of certain gait parameters, such as symmetry, may offer additional information about the practicality of these insoles for certain populations. Although the primary focus of this research was examining the kinematics related to the chain, the symmetry between left and right limb movement may have been affected by insole use. Mean sagittal *knee*, *pelvis* and *global trunk* angle, as well as maximum *global trunk* flexion and *shank* lateral bend angle, all displayed differences between sides post-trial that were not present during the pre-trial. As Gunderson et al. (1989) described how statistical differences between left and right limb angles during phases of gait can be interpreted as asymmetry, it appears the insoles had a negative effect on kinematic symmetry. Patterson et al. (2010) described the importance of symmetry in assessing certain aspects of gait, such as inefficiencies and balance control. The procedure of this study was designed to determine the effects of insole use on the kinematic chain and it is important that further analyses be conducted. Temporal and spatial measures, such as step length, swing time and stance time symmetry ration (Patterson et al., 2010), should be investigated directly before a conclusion is made on the effect of these insoles on symmetry.

6.3 Changes in EMG

The insoles intervention appeared to have a minimal effect on surface EMG. Only two muscles displayed differences between the initial and final collection. Both of these differences, although determined to be significantly different, were very minimal in magnitude. Considering this, it is unclear whether the increase in activation for these muscles, would be beneficial or harmful. The vastus medialis displayed an increase of ≈6 %MVC following insoles use, a change in magnitude between collections of ≈1 %MVC. Similarly, the female non-compliant group displayed an increase in lumbar erector spinae muscle activation between collection sessions, increasing by 3.15 %MVC. Romkes et al. (2006) and Nigg et al. (2006) discussed the potential benefit of strength training footwear causing increased activation. However, Nigg (2001) also discussed how everyday insoles should act to reduce muscle activation if it is encouraging optimal movement. Even though Barefoot Sciences® Products and Services Inc. describes their product as neuromuscular training insoles, they do recommend the product for everyday use. Considering the magnitude of change in muscle activity following insole use, it is unlikely that the changes in muscle activity would be harmful with continuous use.

Similar to kinematics, the insoles appeared to have an effect on the symmetry of muscle activation. The compliant group displayed an improvement in symmetry for gastrocnemius mean muscle activation, with no statistical difference between sides post-trial. With a difference of ≈6.2% MVC pre-intervention, being reduced to a difference of ≈0.9% MVC post-trial. As healthy individuals have displayed symmetrical muscle activation during gait, single limb deficiencies could negatively affect symmetry (Burnett et al., 2011). An increase in symmetry of muscle activation could be an indicator of the insoles correcting for a limb abnormality, however this type of assessment was not performed and to the participants' knowledge they did not have any abnormalities. In contrast to gastrocnemius, a decrease in symmetry was found for the female non-complaint group's lumbar erector spinae activation. Additionally, examining the supplementary ANOVA, differences were found between the external oblique stance and swing side of the body following insole use. While this was not detected for any other back muscles, Vink and Karssemeijer (1988) discussed the benefit of equal activation for controlling the spine during walking. These differences of roughly 1% and 2% MVC although most likely not functional important, indicated that the insoles are potentially affecting symmetry negatively. McGill (2007) discussed the importance of symmetrical muscle stiffness in creating optimal stiffness. Considering the small magnitude in change, as well as the female non-compliant group's small sample size of four, further investigation is required before a conclusion can be made on the effect of insoles on symmetry. Considering a difference was only displayed for two muscles, in addition to the magnitude of change, the insoles appeared to have a minimal effect on symmetry of muscle activation.

6.4 Limitations

There are several potential limitations in this thesis as a result of the instrumentation required to capture 3D-kinematics and surface EMG. Skin motion artifact is often present during collections using surface markers, as the skin is not fixed to the landmark it is placed over. However, extra caution was taken to ensure markers where adhered as best as possible and any artificial gross movement was addressed during collection. The use of surface motion tracking markers for motion analysis results in an estimation of segment motion, and may not be the same as the movement of the bony structures they are to represent (Cappozzo et al., 1995). As such, this potential limitation was minimized.

Every possible precaution was taken in attempt to reduce the potential errors that can be associated with the collection of surface EMG. Skin electrodes are susceptible to recording the muscle activation of additional muscles in the area then the one intended, also called cross contamination (De Luca, 1997). There are also potential limitations associated with the inter-day reliability of surface EMG, such as electrode placement and the tissue composition, which can alter EMG amplitude (De Luca, 1997). To minimize these concerns, the same investigator followed a strict procedure for land marking and adhering electrodes (and markers). While it is not as reliable as intra-day comparisons, De Araújo et al., (2009) reported inter-day reliability to be good following a seven day break between collections. As the trial period was eight weeks long it is possible that changes in muscle and fat size below the electrode could have had an impact on surface EMG amplitude. In an attempt to reduce the variability of surface EMG, extra emphasis was placed on having consistent electrode placement during all collections, as well as an MVC protocol was established for normalization. Considering the few differences observed between pre- and post-trial collections, it is believed that this did not have an effect on the interpretation of EMG results. Cable sway was another possible limitation, as cables were used to connect the electrodes to the portable unit, as well as the portable unit to the amplifier. As such, excess movement may have resulted in a signal artifact or pulling of electrodes. Low pass filters were applied to minimize the effect of movement artifacts without removing part of the surface EMG spectrum, as recommended by Winter (2009). Likewise, the limitations related to surface EMG and motion capture using skin markers, are generally accepted provided precautionary methods are introduced to reduce the error (Preuss and Popovic, 2010).

It is unknown how generalizable certain findings of this study will be to the entire population. Studies in literature using similar methods have reported similar or smaller populations (Tomaro and Burden, 1993; Nester et al., 2003; Nigg et al., 2005; Murley and Bird, 2006; Saha et al., 2007). Considering the main findings of this study, such as the change in thoracic flexion and vastus medialis activation, were found on the entire population, with no differences between compliance (compliant/non-compliant) or sex (male/female) groups, the sample size did not affect the implications of these findings. The sample size of the non-compliant sex groups specifically, may be too small to generalize these findings. When a side-visit-sex interaction was detected, the population was divided into four groups, with only three-four participants in a sex compliance group. The use of a control group may have improved the application of the findings to the general population; however, the resources were not available to collect an additional population due to the intensive instrumentation and time demand. Multiple footwear intervention studies, using similar instrumentation, were also unable to collect on a control group (Nester et al., 2003; Murley and Bird, 2006).

Due to the size of the laboratory and capture area this study was unable to capture the effects of insoles on gait. As gait is typically defined as steady state walking (Winter, 1995), participants would not be able to enter a steady state in the 8m capture space. Winter (2009) described gait initiation as the coordinated movement when going from a stable balance state to a walking state, which occurs in two steps. Similarly, the termination of gait is even more demanding, requiring the forward momentum of the body during steady state walking to be removed within two steps (Winter, 2009). In an attempt to capture walking and not initiation and termination, the first two and last two steps of each trial were removed from analysis, leaving roughly three strides available for comparison. Gait initiation and termination were not examined in this study as it is considered a separate task compared to steady state walking (Winter 1995 and2009). All attempts were made to remove the aspects of locomotion not found in steady state walking, with studies in literature observing a similar number of strides (Reischl et al., 1997; Bird et al., 2003), when evaluating 3D kinematics and muscle activation during walking.

Reexamination of Hypotheses

This study addressed whether an eight week trial period, using a neuromuscular training insole, could alter the 3D joint kinematics and muscle activation during walking. For 3D joint kinematics, the differences in mean, maximum and minimum angle observed illustrated the ability of the insoles to alter walking kinematics. These changes in kinematics however showed no indication of reduced spine flexion as a result of changes in lower limb movement via the kinematic chain. The mean EMG results were not as clear at addressing whether the insoles would impact muscle activation.

Hypothesis #1 Addressed

Hypothesis #1 states: A decrease in internal rotation of the *knee* and *hip*, as well as a decrease in pelvic anterior tilt and spine flexion, will be detected post-trial.

This Hypothesis was REJECTED.

While the insoles did reduce the amount of thoracic flexion exhibited, there were no consistent reductions in *knee* and *hip* internal rotation, *pelvis* anterior tilt, or *lumbar* and *trunk* flexion.

Hypothesis #2 Addressed

Hypothesis #2 states: An increase in mean surface EMG activity will be detected as a result of the neuromuscular training insole.

This hypothesis was REJECTED.

The small change in vastus medialis activation post-trial, although significant, was interpreted to be of little to no physiological consequence. Meaning there was no functional change in muscle activation following insole use.

Hypothesis #3 Addressed

Hypothesis #3 states: The insoles will cause a decrease in mean joint angle, as a result of the increased mean muscle activity of muscles responsible for moving, or stabilizing, said joint.

This hypothesis was REJECTED

No change in both, mean joint angle and mean EMG for muscle crossing said joint, was detected post-trial.

Impact

The interpretations of this study did appear to be of clinical importance regarding the effectiveness of insoles. Even though a reduction in spinal flexion occurred following insoles use, there was no evidence to support that this was result of a reduction in lower limb movements. The very minimal changes in surface EMG indicated that insole use appeared to have little to no functional impact. This however does not act as evidence for the argument that insoles cannot change muscle activation at all. In addition to these findings, the post-trial differences found between and left and right limb kinematics and EMG indicated the need for additional gait parameters, such as symmetry, are examined before making a definitive conclusion about the potential benefits of insoles.

CHAPTER 7

Conclusion

Conclusion

Following an eight week trial period, insoles potentially reduced the amount of spine loading experienced during walking as a result of a reduction in *thoracic spine* movement. The decrease in *thoracic spine* mean, maximum and minimum flexion angles, appear to suggest reduction in the flexion-bending moment experienced in the thoracolumbar junction during walking. This reduction appears to be more significant in females as a decrease in *thoracic spine* range and maximum lateral bend was also found post-trial. However, this reduction in spine motion experienced post-trial was not a result of a reduction in lower limb movements as predicted. This highlights the need for additional theories, besides the kinematic chain, to be explored regarding the use of insoles to beneficially alter spine biomechanics. Another surprising observation was that there was no corresponding change in lumbar thoracic erector spinae activation accompanying the reduction in *thoracic spine* flexion. Some changes were observed in mean surface EMG, particularly a slight increase in vastus medialis activation, nevertheless, these changes are not likely large enough to be functionally relevant in the tested population. While it was not intended, these results highlight the need for studies examining insole use to treat LBP to examine their effects on additional gait parameters such as symmetry. It is important we continue to search for an explanation behind the use of insoles to improve spine biomechanics before we continue to encourage their use. Considering the potential decrease in thoracic loading as a result of reduced spinal flexion, and the little negative side effects found with insole use, it appears appropriate to recommend this product in attempt to reduce LBP.

CHAPTER 8

Knowledge Generated

Knowledge Generated

This study was the first to look at surface EMG for muscles of the back, abdomen and lower limbs, as well as the kinematics pertaining to those muscles, following an insole trial period. While studies in the literature have examined kinematics and surface EMG, most tend to focus on a few joints and muscles. This study was also the first to quantify the effects of insole use on the *trunk*, *lumbar* and *thoracic spine* angle. Previous work in the literature has typically drawn on additional research to connect the lower limbs, pelvis and torso. That is why this thesis intended to create a unique description of how insoles interact with the lower limbs, and how that is transferred up the body, to the upper levels of the spine. This is vital as researchers attempt to combine these segmented studies in effort to infer how the whole body will react. Considering how insoles have been used to treat LBP for years, it is crucial we discover why they work.

The results of this study indicated that prolonged insole use can act to reduce the flexion-bending moment applied to certain functional spine units. The reduction in the mean, maximum and minimum angle observed appeared to provide sufficient evidence to suggest a reduction in thoracic flexion during walking. Females also displayed a reduction in lateral bend, indicating a further potential for the reduction of the flexion-bending moment. Males conversely exhibited an increase in lateral bend, however the reduction in thoracic flexion appeared to be greater, again suggesting an overall reduction in the flexion-bending moment. While this study provided evidence of decreased thoracic flexion, there are no results indicating why.

This study added more evidence to the literature suggesting that the effects of insoles on *knee* and *hip* kinematics are minimal. Likewise this study found no evidence to support the idea of reduced spine flexion as a result of the kinematic chain. While it is possible that changes in the motion of one joint can alter the motion of joints above and below it, there is no indication that the magnitude of these changes during walking are large enough to cause this to occur. It is evident that there is a need for additional theories behind the success of insoles to treat LBP if we are to guarantee there is more than a placebo effect occurring.

The surface EMG results add to the evidence suggesting it is possible for insole use to alter EMG. It is yet to be determined if these changes are of functional importance. Similar to previous studies, it also adds more evidence suggesting it is unclear what direction muscle activity will change. Whether increasing muscle activation over a prolonged period is beneficial, or harmful, must still be determined, as there is more to consider when investigating their impact on muscle activity.

Multiple studies in the literature have examined the effects of insoles relating to the kinematic chain, though none of these studies have examined their effect on kinematic symmetry. Although it was not the primary focus of this study to determine the effect of insoles on symmetry, it was quite evident that insoles appeared to contribute to differences between left and right limb motion post-trial. While symmetry may not be a primary focus of biomechanists investigating insoles in the past, it should be taken into consideration when evaluating the effects of insoles on walking.

CHAPTER 9

Future Directions

Future Directions

Considering the multiple significant difference found when only examining kinematics, or EMG, and not both, it is important to understand the immediate and time-varying (i.e. weekly movement quantification vs. pre/post 8 weeks of use) effects of the insoles. The most extreme example of this was how no changes where exhibited for lumbar erector spinae activation, even though multiple variables suggested participants were standing more upright. The immediate and time-varying effects of these insoles must also be quantified in order to determine if the changes in joint kinematics were a result of an increase in muscle activation. This would also provide evidence to suggest whether or not the muscles are growing stronger with insole use. It is not enough to know that insoles can reduce spine flexion, we must determine the means behind this reduction.

While this study observed kinematic angle during stance phase of walking, it is important that we test the effects of insoles on additional gait parameters, as well as static, and more physical demanding dynamic tasks. The results suggesting a negative impact on symmetry highlight the importance of factors other than spine loading being examined. If an insole is to be used for everyday use, it is important we understand the effects of an insole on everyday tasks. If the insoles have a similar effect on standing, they could serve to reduce the flexion-bending moment during most aspects of everyday life. In addition to this, intense physical activity is also known to be more demanding on the body and examination of these tasks potentially could reveal a larger impact. Insoles acting to reduce joint loading such as running, jumping, or other intense exercises could also be of great value in helping to reduce LBP.

This study highlighted the idea that different people respond differently to a specific insole. In order to determine the effectiveness of a design, participants whose foot type it is intended to complement, or correct, should be the primary focus of the initial research. Investigating the effects of an insole on a healthy population is ill-advised initially, as it is hindering the ability of the research to detect if they can work at all. In addition to this, participants who have no reported history of orthotic use or back pain may be less likely to use the insoles as much as those with a history of foot and back problems. Once shown that the insoles work on the intended population, then expanding the population may be beneficial. Research examining the effects of an insole on kinematics adding foot type to its selection criteria appears appropriate.

CHAPTER 10

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Appendices

Appendix A- Kinematic Variables Pre-Post trial period

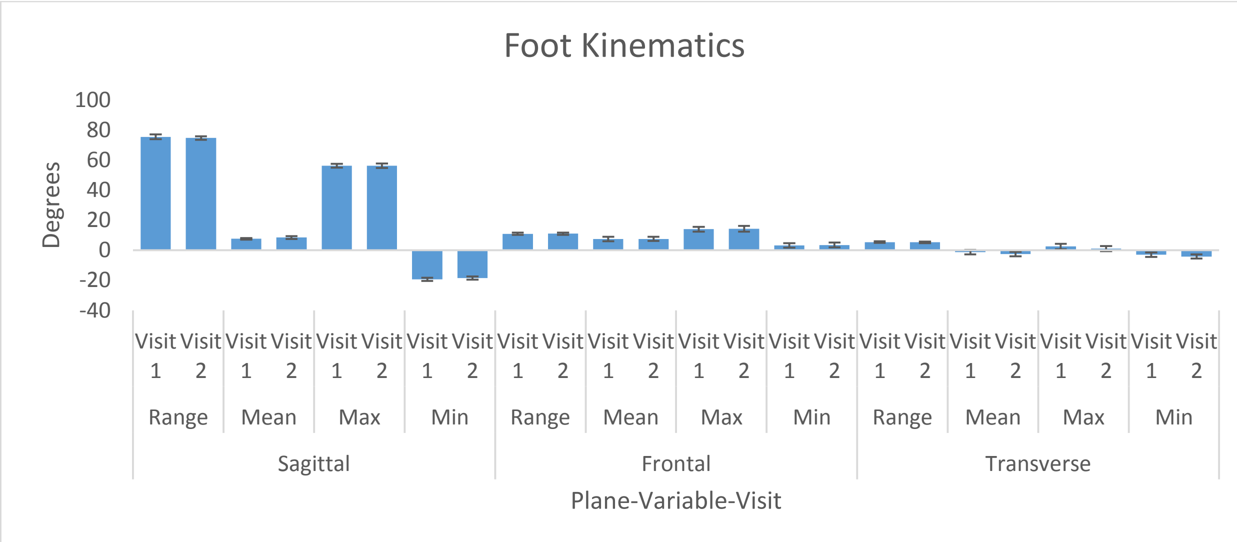

Figure A1: Foot segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

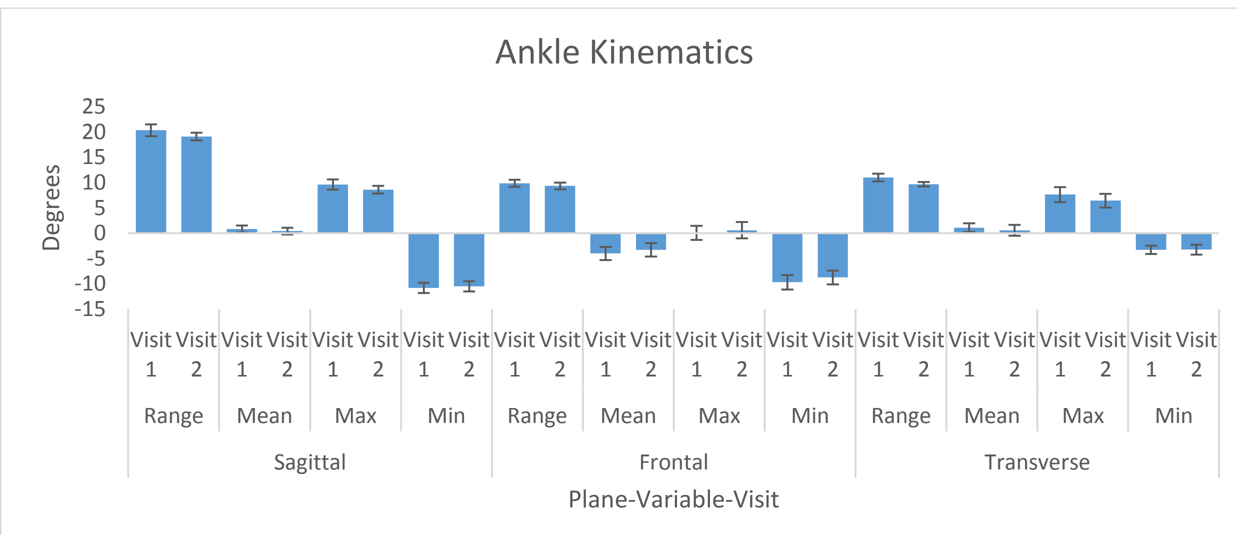

Figure A2: Ankle segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

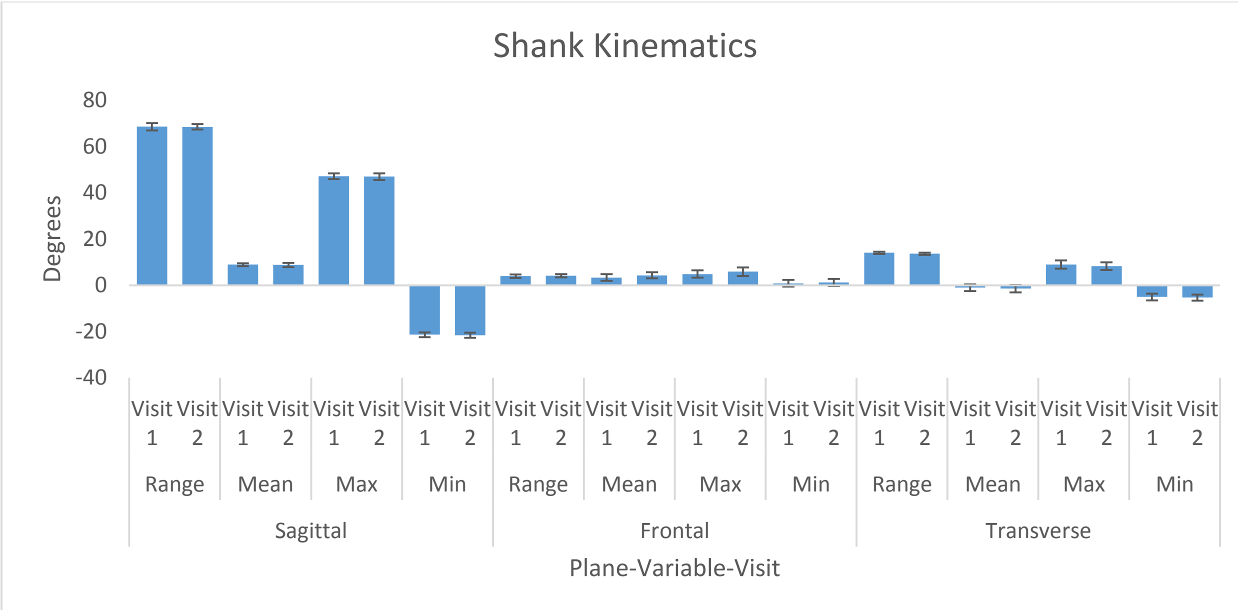

Figure A3: Shank segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

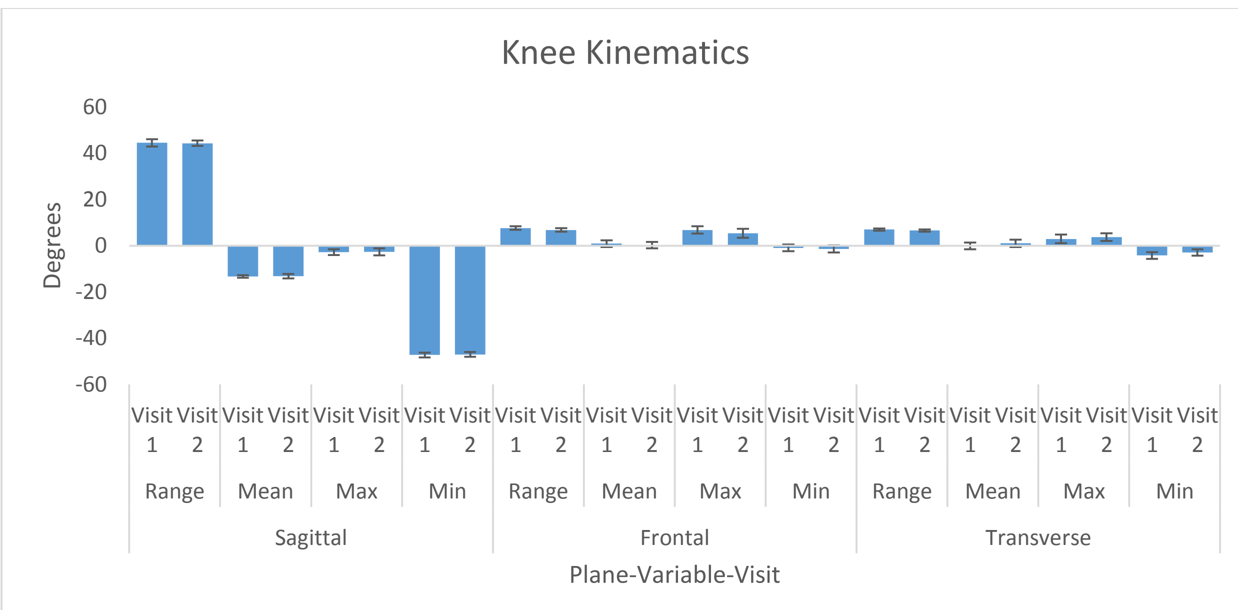

Figure A4: Knee segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

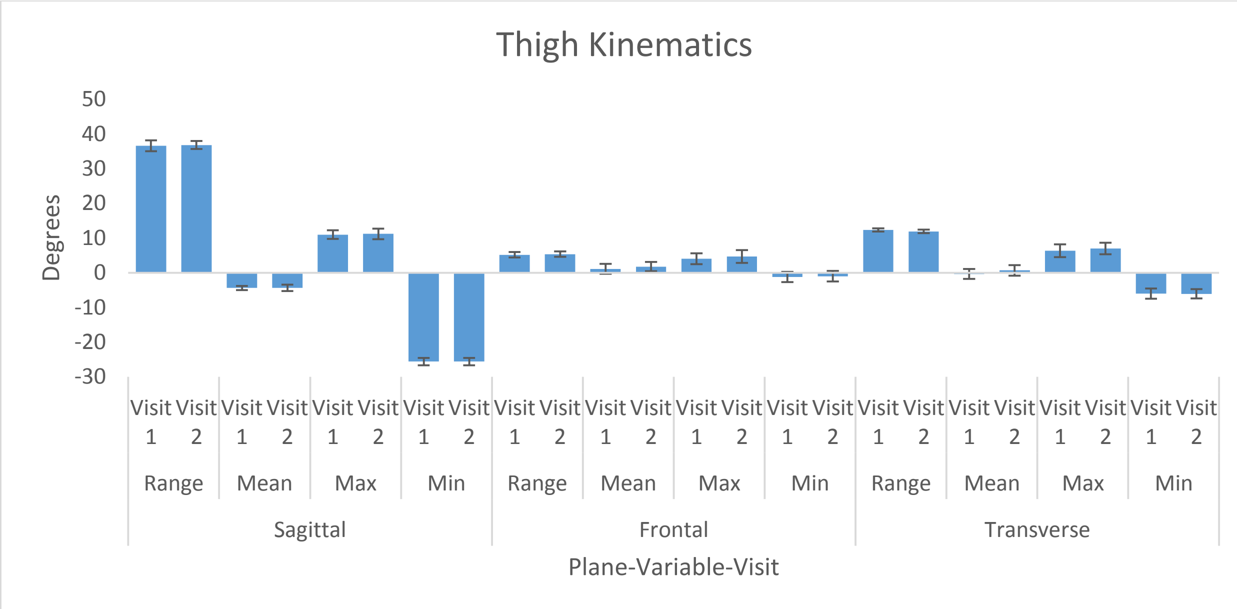

Figure A5: Thigh segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

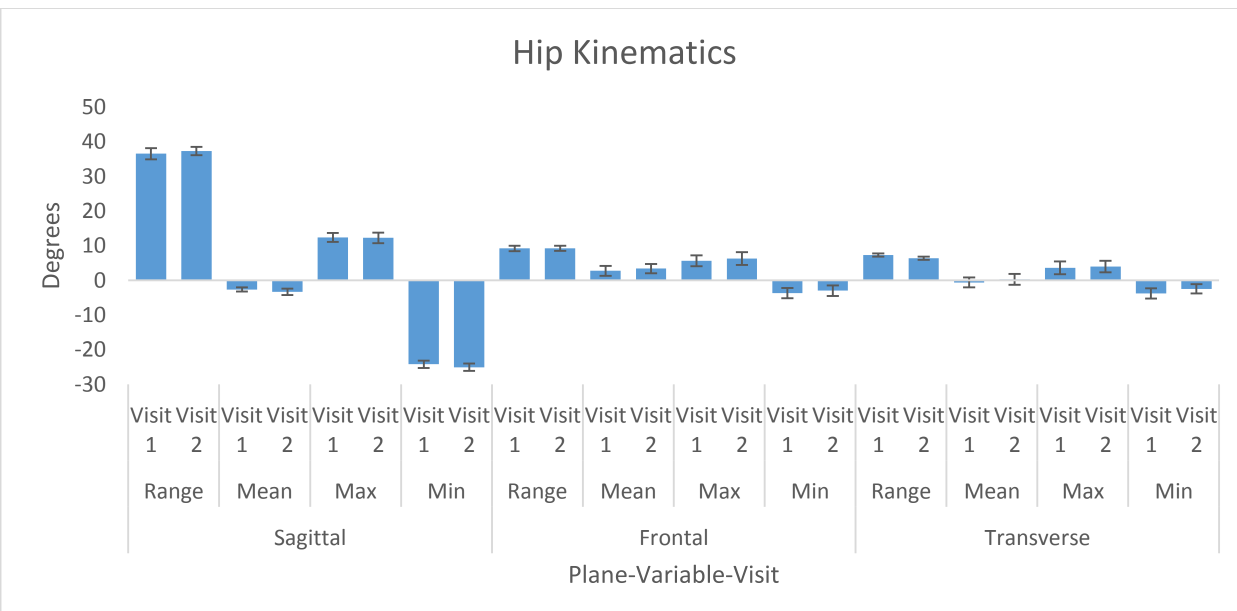

Figure A6: Hip segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

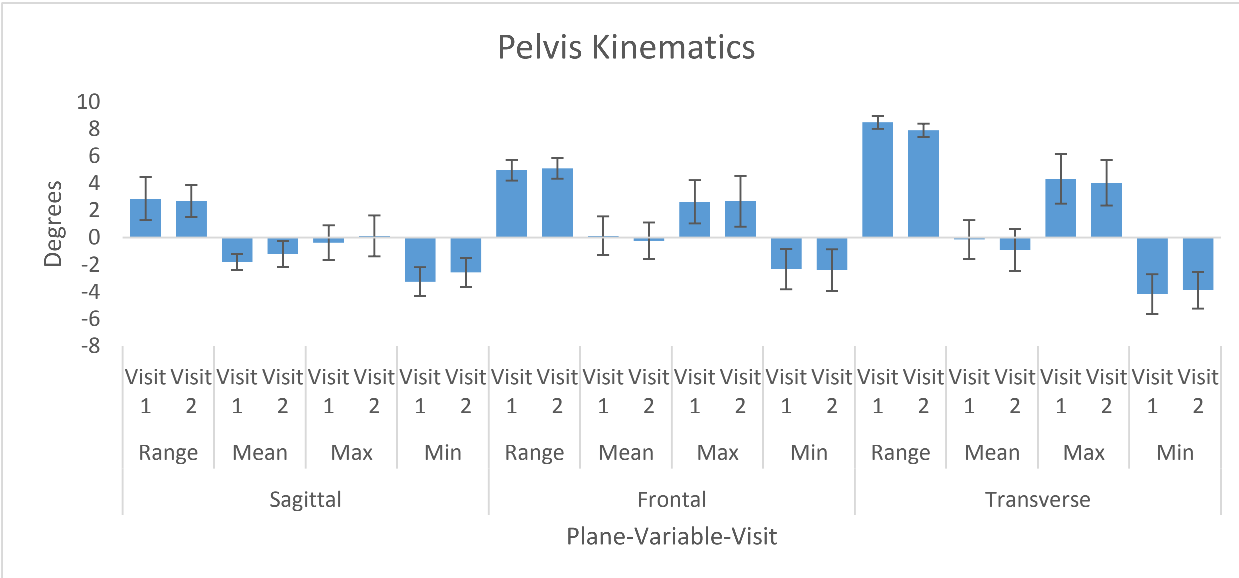

Figure A7: Pelvis segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

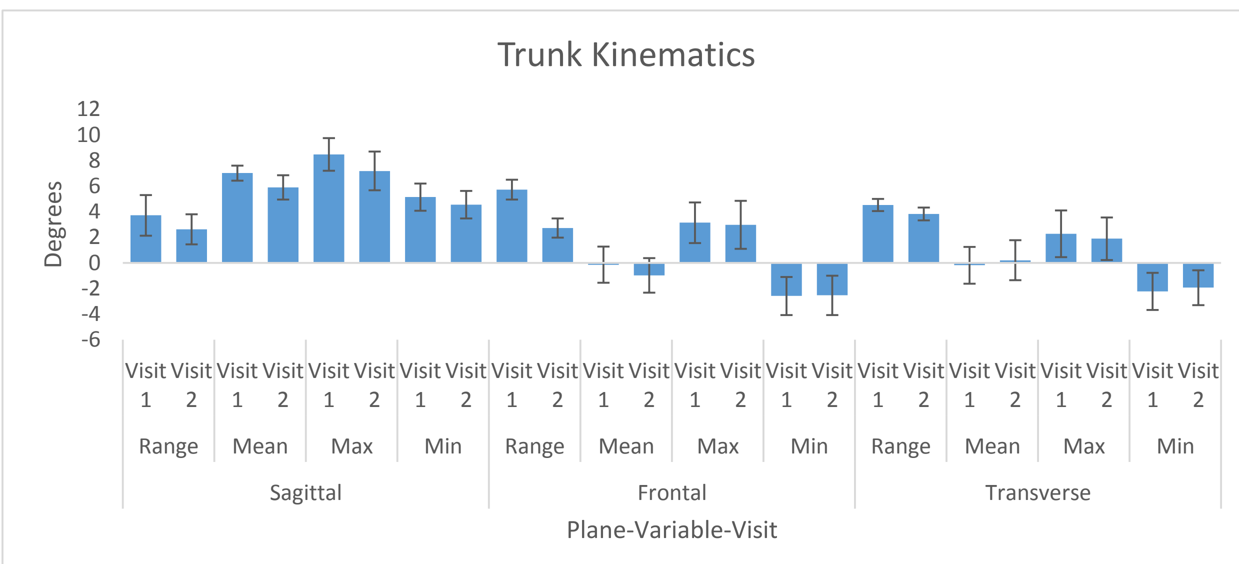

Figure A8: Trunk segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

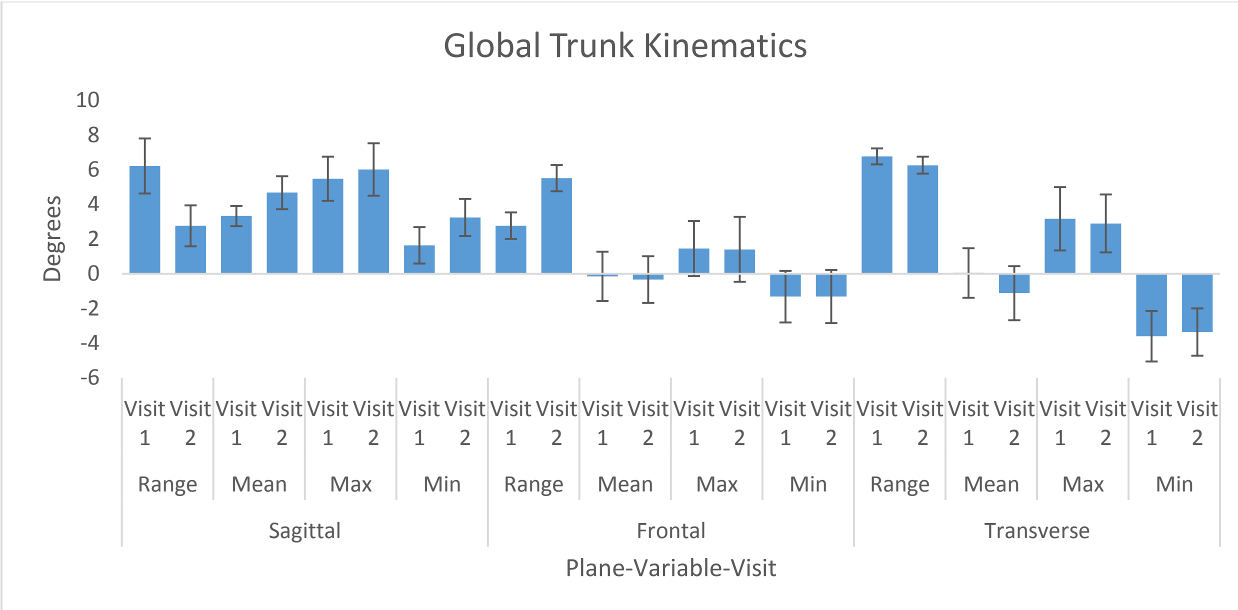

Figure A9: Global Trunk segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

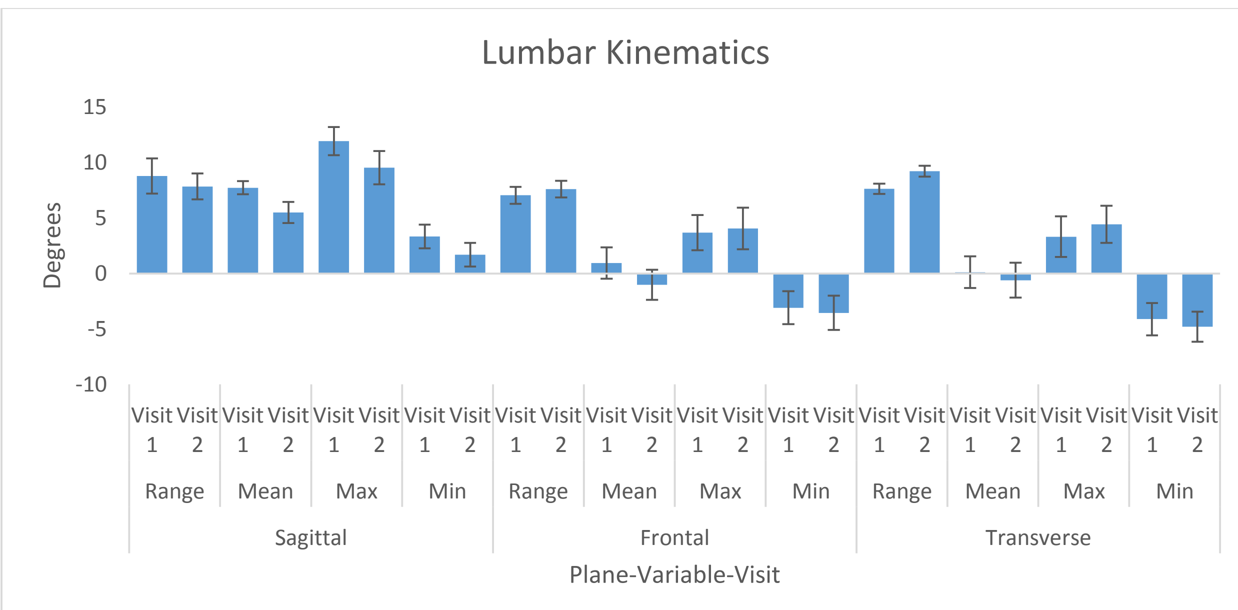

Figure A10: Lumbar segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

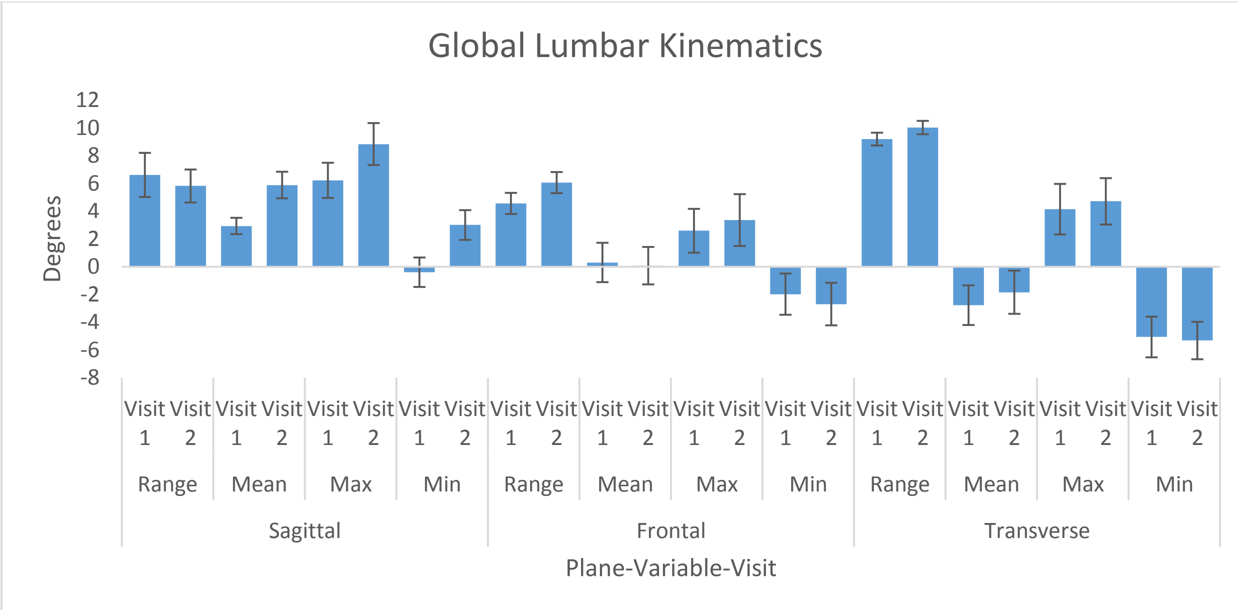

Figure A11: Global Lumbar segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

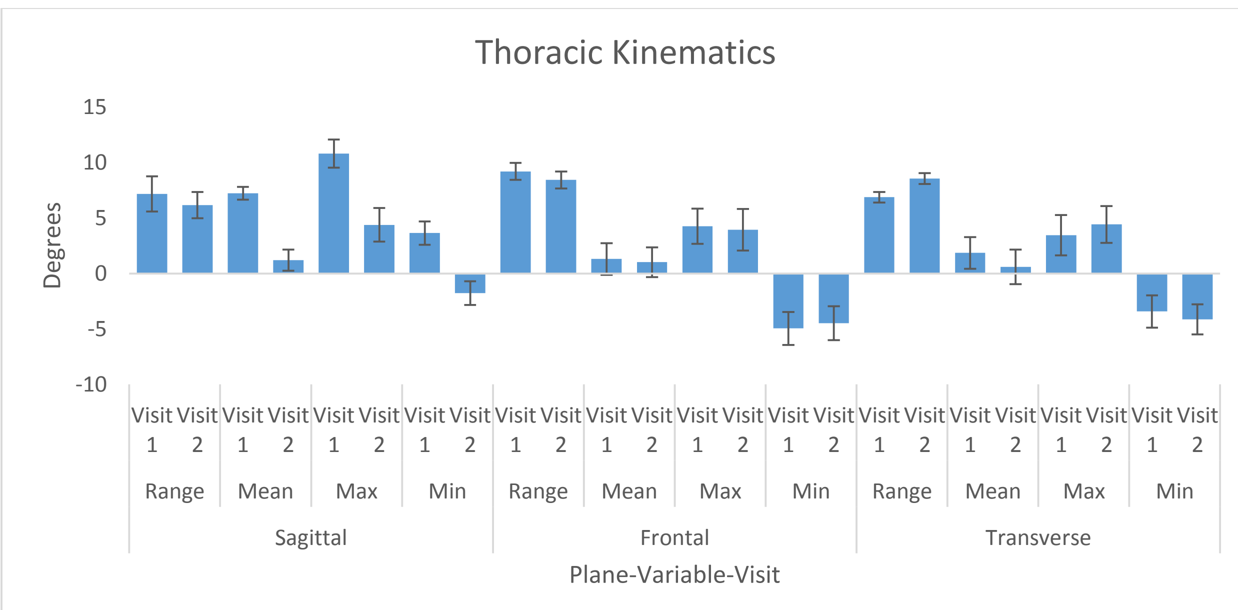

Figure A12: Thoracic segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

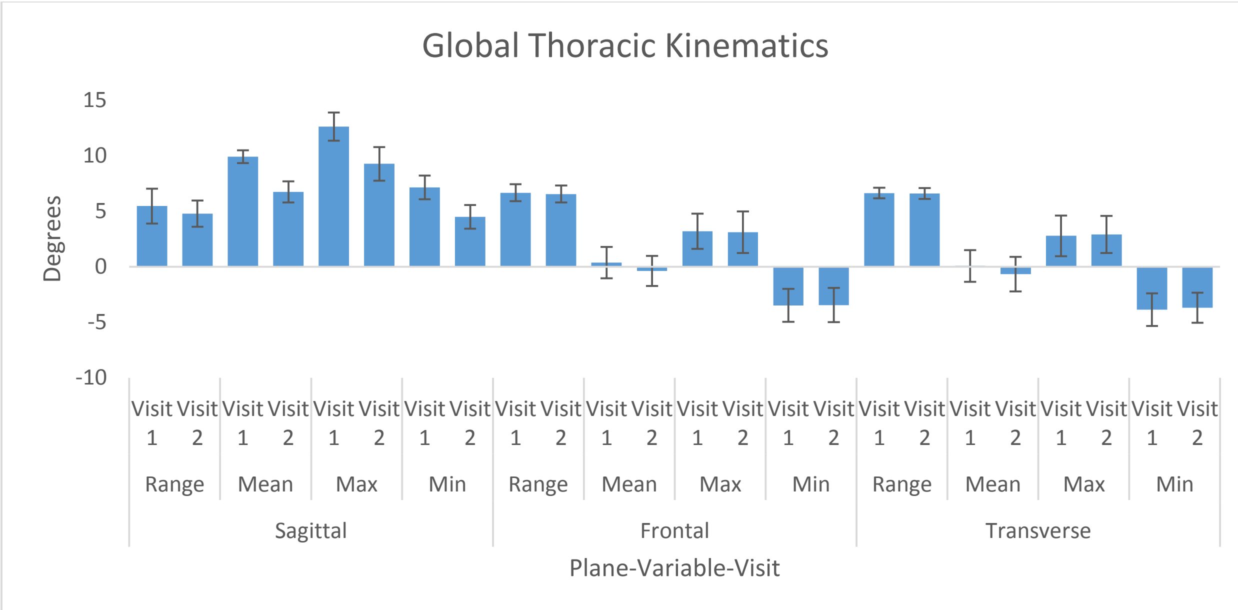

Figure A13: Global Thoracic segment range of motion (range), mean, maximum and minimum angle, for each plane ±SEM

Appendix B- Muscle Activation Pre-Post trial period

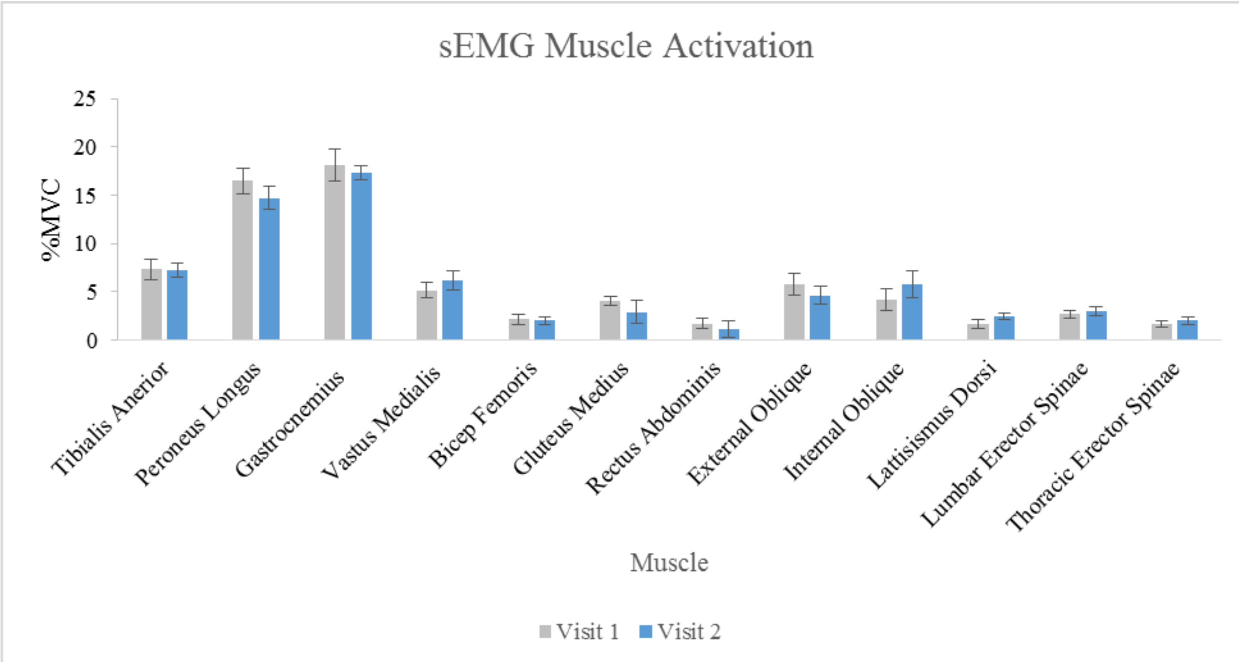

Figure B.1: Indicates participants' muscle activation as %MVC (±SEM) during the stance phase of walking.

Appendix C- Statistical Summary Tables

Table C1: Display all MANOVA main effects and interactions for mean surface EMG and mean joint angle in the sagittal plane. Highlighted yellow indicates a significant interaction or main effect at $p<0.05$ involving the factor of visit.

MANOVA Mean EMG + X Joint Angle (%MVC)	
Muscle-Joint	Interaction/Main effects
Tibialis Anterior-Ankle	-Sex (p=0.003) -Side*level (p=0.029)
Peroneous Longus-Ankle	-Sex (p=0.027)
Gastrocnemius-Ankle	-Side*level (p=0.003)
Bicep Femoris-Knee	-Level (p=0.009) -Sex (p=0.025)
Vastus Medialis-Knee	-Level (p=0.001)
Gluteus Medius-Hip	-level (p=0.018)
External Obliques-Trunk	-Side*sex*level (p=0.044)
Internal Obliques-Trunk	-Sex (p=0.046)
Rectus Abdominus-Trunk	-None
Latissimus Dorsi-Trunk	-None
Lower Thoracic Erector spine-Thoracic spine	-Side*sex*level (p=0.042)
Lumbar Erector Spine-Lumbar spine	- Side*visit*sex*level (p=0.035)

Table C2: Display all MANOVA main effects and interactions for mean surface EMG and mean joint angle in the frontal plane. Highlighted yellow indicates a significant interaction or main effect at p <0.05 involving the factor of visit.

MANOVA Mean EMG + Y Joint Angle (%MVC)	
Muscle-Joint	Interaction/Main effects
Tibialis Anterior-Ankle	-Sex (p=0.005)
Peroneous Longus-Ankle	-Sex (p=0.045)
Gastrocnemius-Ankle	None
Bicep Femoris-Knee	-Sex (p=0.015)
Vastus Medialis-Knee	None
Gluteus Medius-Hip	-None
External Obliques-Trunk	-Sex (p=0.010) -Side (p=0.005)
Internal Obliques-Trunk	-Side (p=0.033)
Rectus Abdominus-Trunk	-Side (p=0.038)
Latissimus Dorsi-Trunk	-Side (p=0.038)
Lower Thoracic Erector spine-Thoracic spine	-None
Lumbar Erector Spine-Lumbar spine	-Side*visit*sex*level (p=0.040)

Table C3: Display all MANOVA main effects and interactions for mean surface EMG and mean joint angle in the transverse plane. Highlighted yellow indicates a significant interaction or main effect at p <0.05 involving

MANOVA Mean EMG + Z Joint Angle (%MVC)	
Muscle-Joint	Interaction/Main effects
Tibialis Anterior-Ankle	-Sex (p=0.005)
Peroneous Longus-Ankle	-Sex (p=0.044)
Gastrocnemius-Ankle	-None
Bicep Femoris-Knee	-Sex (p=0.021) -Side (p=0.020)
Vastus Medialis-Knee	-Side (p=0.017)
Gluteus Medius-Hip	-None
External Obliques-Trunk	-Sex (p=0.006) -Side*level (p=.035)
Internal Obliques-Trunk	-Sex (p=.049)
Rectus Abdominus-Trunk	-None
Latissimus Dorsi-Trunk	-None
Lower Thoracic Erector spine-Thoracic spine	-None
Lumbar Erector Spine-Lumbar spine	- Sex (p=.034)

Tables C4-9 display all main effects of a four-way mixed ANOVA) run on the range of motion as well as maximum, minimum, mean angle observed, for each plane separately. The two repeated factors used in the analysis were insole condition (visit 1, visit 2) and side (left-right limb stance phase) and the two between group factors were sex and compliance. Males were represented by M and females by F. Yellow indicates interactions or main effects involving visit and green indicates a difference between left and right limb stance that is only present for one visit. * Indicates a significance difference of *p* <0.05. The letters NC and C display the values for the non-compliant and compliant group respectively. The letters L,v,s,sx,st next to * indicate whether the factors of level, visit, side, sex and/or step were detected in the interaction or main effect.

Tables C4: Maximum and minimum sagittal plane angle (X)

		MAX X				MIN X			
Angle	Sex	Visit 1		Visit 2		Visit 1		Visit 2	
		Left	Right	Left	Right	Left	Right	Left	Right
Ankle	Male	9.63 (1.02)(*p*=.29)		8.61 (0.77)		-11.92 (1.50) *s(*p*=.042)	-9.38 (0.83) *s(*p*=.042)	Collapsed visit	
	F								
Foot	M	56.30 (1.27)(*p*=.95)		56.38 (1.51)		-16.03 (1.07) *sx(*p*=.001)		Collapsed visit	
	F					-21.58(1.16) *sx(*p*=.001)			
Shank	M	47.17 (0.79)(*p*=.20)		47.00 (0.89)		-20.91 (0.71) *s(*p*=.006)	-22.36 (0.61) *s(*p*=.006)	Collapsed visit	
	F								
Shank (Level)						NC -20.44(0.68) *L(*p*=.029) C -22.83(0.85) *L(*p*=.029)		Collapsed visit	
Knee	M	-2.81 (1.03)(*p*=.85)		-2.70 (1.01)		-46.86 (1.27)		-48.12 (1.89)	
	F					-47.14 (2.05) *v(*p*=.036)		-45.25 (1.41) *v(*p*=.036)	
Thigh	M	11.00 (0.76)(*p*=.53)		11.24 (0.67)		-24.89 (0.60)	-26.21 (0.91)	-26.01 (0.72)	-24.95 (0.88)
	F								
Hip	M	12.38 (1.01)(*p*=.89)		12.27 (0.95)		NC -23.92 (1.55) *v(*p*=.023) C -24.49 (1.75)		NC -26.65 (1.39) *v(*p*=.023) C -22.87 (1.58)	
	F								
Pelvis	M	-0.38 (0.71)(*p*=.48)		0.11(0.78)		NC -3.75 (0.91) *v(*p*=.030) C -1.89 (0.93)		NC -2.59 (1.03) *v(*p*=.030) C -3.45 (1.06)	
	F								
Trunk	M	8.47 (1.04)(*p*=.11)		7.18 (1.13)		5.14 (1.22)		4.55 (1.12)	
	F								

Trunk Global	M	5.29 (1.60)	5.02 (1.48)	6.17 (0.87) *s(*p*=.001)	5.56 (0.85) *s(*p*=.001)	2.56 (1.68) *s(*p*=.018)	2.33 (1.68) *s(*p*=.018)	Collapsed visit
	F							
Lumbar	M	11.93 (3.84)(*p*=.54)		9.53 (1.54)		3.34 (3.53)		1.69 (1.65)
	F							
Thoracic	M	10.81 (2.56) *v(*p*=.004)		4.38 (2.53) *v(*p*=.004)		3.64 (2.59) *v(*p*=.019)		-1.79 (2.31) *v(*p*=.019)
	F							
Global Lumbar	M	5.71 (2.06)	6.76 (2.20)	9.09 (1.54)	8.60 (1.61)	-0.39 (2.14)		3.02 (1.64)
	F							
Global Thoracic	M	12.63 (1.67)		9.29 (1.60)		7.16 (1.53)		4.50 (1.66)
	F							

Tables C5: Mean and range of motion for the sagittal plane (X)

		Mean X				Range X			
Angle	Sex	Visit 1		Visit 2		Visit 1		Visit 2	
		Left	Right	Left	Right	Left	Right	Left	Right
Ankle	M F	NC1.94 (0.97) *L(p=.013) C-1.12 (1.40) *L(p=.013)	NC 1.00(0.83) C 0.14 (0.73)	Collapsed visit		21.43 (1.61)* s(p=.037)	18.03 (0.67)* s(p=.037)	Collapsed visit	
Foot	M	8.95 (0.93) *s(p=.007)	7.25 (0.78) *s(p=.007)	Collapsed Visit		72.42 (0.95) *sx(p=.027)		Collapsed visit	
	F					77.86 (2.24) *sx(p=.027)			
Shank	M F	NC 11.00 (0.61)* l(p=.036) s(P=0.013) C 7.89 (1.25)* l(p=.036)	NC 9.23 (0.89)* v(p=0.047) s(P=0.003) C 6.95 (0.89)	NC 11.04 (0.95)* s(p=0.001) C 8.15 (1.22)	NC 7.87 (0.94) * v(p=0.047) s(p=0.001) C 7.75 (0.78)	68.62 (0.79)		68.58 (0.82)	
Knee	M	-12.41 (1.35)	-11.65 (1.65)	-13.01 (1.99)	-12.45 (1.58)	44.52 (0.74)		44.32 (0.94)	
	F	-13.88 (2.00)	-13.93 (1.92)* v(p=.014)	-14.98 (1.84)* s(p=0.008)	-11.19 (0.95)* v(p=.014) s(p=.008)				
Knee (level)	M F	NC -15.23 (1.31) *l(p=.030) C-10.60 (1.23) *l(p=.030)				Collapsed visit			
Thigh	M F	NC-5.42(0.61) *l(p=.005) C-3.03(0.44) *l(p=.005)		Collapsed visit		36.62 (0.87)		36.84 (0.95)	
Hip	M F	NC-3.15 (1.48) *v(p=.044) C -1.98 (1.10		NC -5.30 (0.71) * v(p=.044),l(p=.011) C -0.74 (1.53)*l(p=.011)		36.53 (0.93)		37.30 (0.99)	
Hip (Side*Visit)	M F	-2.14 (1.01)	-2.99 (1.03)	-3.46 (1.08)	-2.58 (0.91)				
Pelvis	M F	NC-2.19(0.96) *v(p=.038) C-1.34(0.98)		NC-0.43(0.79) *v(p=.038) C-2.23(1.28)		2.86 (0.25)		2.68 (0.19)	
Pelvis (Side*Visit)	M F	-1.77(0.68)	- 1.76(0.67)	-1.15 (- 0.75) * s(p=.005)	-1.51 (0.72) * s(p=.005)				
Trunk	M F	7.02 (1.03)		5.90 (1.11)		3.71 (0.81)		2.63 (0.23)	
Trunk Global	M	3.24 (2.38)	3.26 (2.17)	4.77 (0.86) *	4.33(0.83) *	6.22 (2.27)		2.77 (0.15)	

	F			s(p=.001)	s(p=.001)		
Lumbar	M	7.73 (3.64)		5.49 (1.60)		7.85 (0.54)	7.63 (0.58)
	F						
Thoracic	M F	7.23 (2.55) * v(p=.004)		1.20 (2.37) * v(p=.004)		6.75 (0.57)	5.81 (0.63)
Global Lumbar	M	2.94 (2.11)		5.90 (1.53)		6.62 (0.48)	5.83 (0.51)
	F						
Global Thoracic	M	9.92 (1.60)		6.74 (1.73)		5.47 (0.49)	4.79 (0.52)
	F						

Tables C6: Maximum and minimum frontal plane angle (Y)

		MAX Y				MIN Y			
Angle	Sex	Visit 1		Visit 2		Visit 1		Visit 2	
		Left	Right	Left	Right	Left	Right	Left	Right
Ankle	M	0.07 (1.40)		0.60 (1.61)		-9.70 (1.44)		-8.74 (1.36)	
	F								
Foot	M	14.09 (1.59)		14.30 (1.87)		3.20 (1.48)		3.23 (1.45)	
	F								
Shank	M	4.71 (0.61)	5.02 (0.45)* v(p=.014)	5.32 (0.70)*s(p=.012)	6.49 (0.49)* v(p=.014) s(p=.012)	0.88 (0.51)		1.75 (0.58)	
	F								
Knee	M	6.83 (0.87)		5.38 (1.25)		-0.94 (0.36)		-1.43 (0.42)	
	F								
Thigh	M	2.79 (0.32)		3.24 (0.26)		-1.18 (0.40)* s(p=.031)	-0.49 (0.50)* s(p=.031)	Collapsed visit	
	F								
Hip	M	5.10 (0.69) *sx(p=.016)				-2.04 (0.50) *sx(p=.001)			
	F	6.84 (0.74) *sx(p=.016)				-4.57 (0.55) *sx(p=.001)			
Hip (side)	M					-3.83 (0.56) *s(p=.032)	-2.78 (0.54) *s(p=.032)	Collapsed visit	
	F								
Pelvis	M	1.44 (0.22) *sx(p=.001)				-1.64 (0.53) *sx(p=.001)			
	F	3.30 (0.31) *sx(p=.001)				-3.67 (0.68) *sx(p=.001)			
Trunk	M	3.14 (0.52)		2.98 (0.41)		-2.17 (0.28) *sx(p=.046)			
	F					-2.95 (0.29) *sx(p=.046)			
Trunk Global	M	1.45 (0.38)		1.40 (0.35)		-1.32 (0.12)		-1.31 (0.15)	
	F								
Lumbar	M	4.80 *sx(p=.004)				-3.09 (0.23)		-3.55 (0.25)	
	F	2.91 *sx(p=.004)							
Thoracic	M	3.38 (1.02)* v(p=.005)		4.63 (2.00) * v(p=.005)		-3.24 (0.48)		-3.07 (0.49)	
	F	5.11 (1.65) * v(p=.042)		3.26 (1.36) * v(p=.042)					
Global Lumbar	M	2.60 (0.24)		3.38 (0.55)		-1.98 (0.20)		-2.69 (0.48)	
	F								
Global Thoracic	M	3.21 (0.25)		3.12 (0.27)		-3.47 (0.23)		-3.44 (0.28)	
	F								

Tables C7: Mean and range of motion for the frontal plane (Y)

		Mean Y				Range Y			
Angle	Sex	Visit 1		Visit 2		Visit 1		Visit 2	
		Left	Right	Left	Right	Left	Right	Left	Right
Ankle	M	-4.02 (1.28)		-3.30 (1.33)		8.38 (0.58) *sx(*p=.034)*			
	F					10.81 (1.07) * sx(*p=.034)*			
Foot	M	7.51 (1.42)		7.62 (1.35)		9.68 (0.60) *sx(*p*=.039)			
	F					12.40 (1.24) *sx(*p*=.039)			
Shank	M	3.34 (0.45)		4.35 (0.57)		3.99 (0.18)		4.16 (0.18)	
	F								
Knee	M	1.38 (0.46)* v(*p*=0.17)	0.76 (0.50)	-1.14 (0.74)* v(*p*=.017) sx(*p=.033*	0.13 (0.57)	7.62 (0.61)		6.82 (0.88)	
	F	0.87 (0.63)	0.66 (0.49)	1.73 (0.96)* sx(*p=.033)*	0.37 (0.44)				
Thigh	M	1.12 (0.41)		1.76 (0.27)		5.20 (0.37)		5.38 (0.25)	
	F								
Hip	M	2.76 (0.38)		3.42 (0.35)		NC 10.02 (1.01) * s(*p*=.001) C 8.91 (1.00)	NC 8.87 (0.88) * s(*p*=.001) C 9.02 (1.05)	Collapsed visit	
	F								
Pelvis	M	-0.23 (0.41) *s(*p*=.016)	0.52 (0.40) *s(*p*=.016)	Collapsed visit		3.08 (0.42) *sx(*p*=.001)			
	F	-1.01 (0.45) *s(*p*=.001)	0.93 (0.49) *s(*p*=.001)			6.97 (0.62) *sx(*p*=.001)			
Trunk	M	-1.09 (0.51)* s(*p*=001)	0.805 (0.35)* v(*p*=.009)	-1.14 (0.36)	-0.83 (0.28) * v(*p*=.009)	NC 10.02 (0.68) C 8.91 (0.74)	NC 8.87 (0.68) C 9.02 (0.74)	Collapsed visit	
	F								

			s(p=.001)					
Trunk (sex)	M					4.46 (0.59) *sx(p=.041)		
	F					6.46 (0.64) *sx(p=.041)		
Trunk Global	M	-0.40 (0.34)* S(p=.001)	0.66 (0.25)* sx(p=.020) s(p=.001)	Collapsed visit		2.82 (0.45)		3.07 (0.55)
	F	-0.69 (0.63)	-0.54 (0.50)* sx(p=.020)			2.73 (0.30) *v (p=.047)		2.34 (0.30) *v (p=.047)
Lumbar	M	-1.05 (1.26)* S(p=.001)	0.95 (1.13)* S(p=.001)	Collapsed visit		4.58 (0.43)		6.07 (1.03)
	F							
Thoracic	M	1.20 (0.88)		1.01 (1.22)		7.41 (1.12) *v(p=.003)		9.88 (1.54) *v(p=.003)
	F					10.99 (1.32) *v(p=.045)		6.98 (0.87) *v(p=.045)
Global Lumbar	M	-0.70 * s(p=.001)	1.09 * s(p=.001)	Collapsed visit		4.58 (0.43)		6.07 (1.03)
	F							
Global Thoracic	M	0.66 * s(p=.001)	-0.46 * s(p=.001)	Collapsed visit		6.68 (0.46)		6.57 (0.53)
	F							

Tables C8: Maximum and minimum transvers plane angle (Z)

		MAX Z				MIN Z			
Angle	Sex	Visit 1		Visit 2		Visit 1		Visit 2	
		Left	Right	Left	Right	Left	Right	Left	Right
Ankle	M	7.63 (1.48)		6.42 (1.34)		-3.31 (0.83)		-3.24 (0.98)	
	F								
Foot	M	5.41 (2.05) *sx(*p*=.009)				-6.49 (1.96) *sx(*p*=.031)			
	F	-1.72 (2.18) *sx(*p*=.009)				-0.66 (1.49) *sx(*p*=.031)			
Shank	M	9.00 (1.67)		8.34 (1.87)		-5.03 (1.02)		-5.36 (1.21)	
	F								
Knee	M	2.94 (1.61)		3.68 (0.89)		-4.23 (2.22)		-2.92 (0.83)	
	F								
Thigh	M	3.87 *sx(*p*=.007)				-6.00 (1.66)		-4.91 (0.87)	
	F	9.52 *sx(*p*=.007)							
Hip	M	1.38 (1.31) *sx(*p*=.005)				-3.77 (1.62)		-2.42 (0.95)	
	F	6.23 (1.95) *sx(*p*=.005)							
Pelvis	M	4.17 (0.49)		3.87 (0.49)		-4.32 (0.81)		-4.03 (0.79)	
	F								
Trunk	M F	NC 3.15 (0.63)* v(*p*=.042) C 1.54 (0.57)	NC 2.13 (0.68) C 2.10 (0.92)	NC 1.77 (0.31)* v(*p*=.042) C 2.36 (0.36)	NC 2.32 (0.46) C 1.00 (0.53)	-2.23 (0.31)		-1.94 (0.20)	
Trunk Global	M	3.17 (0.54)		2.90 (0.83)		-3.61 (0.28)		-3.36 (0.25)	
	F								
Lumbar	M	3.32 (1.92)		4.43 (1.35)		-4.11 (0.44)		-4.79 (0.58)	
	F								
Thoracic	M	3.45 (1.44)		4.42 (1.38)		-1.80 (0.95)		-2.71 (0.57)	
	F								
Global Lumbar	M	4.15 (0.84)		4.73 (0.63)		-2.88 (1.31) *s(*p*=.043)	-7.83 (1.65) *s(*p*=.043)	Collapsed visit	
	F								
Global Thoracic	M	2.79 (0.23)		2.92 (0.18)		-3.86 (0.69)		-3.68 (0.31)	
	F								

Tables C9: Mean and range of motion for the transvers plane (Z)

		Mean Z				Range Z			
		Visit 1		Visit 2		Visit 1		Visit 2	
Angle	Sex	Left	Right	Left	Right	Left	Right	Left	Right
Ankle	M	1.10 (0.88)		0.56 (1.06)		9.30 (0.70) *sx(p=.013)			
	F					11.35 (0.84) *sx(p=.013)			
Foot	M	-4.81 (2.07) *sx(p=.023)				5.52 (0.47)		5.32 (0.49)	
	F	1.13 (1.63) *sx(p=.023)							
Shank	M	-4.03 (1.71) *sx(p=.007)				11.33 (0.64) *sx(p=.001)			
	F	1.55 (0.95) *sx(p=.007)				15.96 (1.38) *sx(p=.001)			
Shank (level)						NC 15.39 (1.34) *l(p=.008) C 11.90 (0.97) *l(p=.008)			
Knee	M	-0.08 (1.55)		1.03 (0.74)		7.03 (1.07)		6.60 (0.50)	
	F								
Thigh	M	-0.33 (1.57)		0.69 (0.96)		9.16 (0.90) *sx(p=.001)			
	F					14.54 (1.67) *sx(p=.001)			
Thigh (level)	M					NC 14.23 *l(p=.002) C 9.47 *l(p=.002)			
	F								
Hip	M	-0.59 (1.23)		0.26 (0.89)		5.27 (0.50) *sx(p=.001)			
	F					8.14 (1.17) *sx(p=.001)			
Hip (level)						NC 7.92 (1.09) *l(p=.003) C 5.49 (0.55) *l(p=.003)			
Pelvis	M	-0.01 (0.67) *s(p=.006)	-1.05 (0.75) *s(p=.006)	Collapsed visit		8.48 (0.92)		7.90 (0.84)	
	F								
Trunk	M	-0.19 (0.35)		0.20 (0.07)		4.52 (0.63)		3.82 (0.38)	
	F								
Trunk Global	M	0.12 (0.58)*s (p=.001)	-1.19 (0.65)* s(p=.001)	Collapsed visit		6.77 (0.52)		6.26 (0.58)	
	F								
Lumbar	M	0.11 (1.65)		-0.60 (1.20)		8.01 (0.88) * s(p=.005)	8.84 (0.85) * s(p=.005)	Collapsed Visit	
	F								
Thoracic	M	2.17 (2.21)	2.93 (2.97)	0.95 (1.87)	1.47 (2.20)	6.96 (0.83)		8.42 (1.24)	
	F	1.38 (1.59)	0.93 (1.58)	-0.64 (1.64)	0.58 (1.23)				
Global Lumbar	M	-1.58 (1.31) *s(p=.001)	-3.97 (1.45) *s(p=.001)	Collapsed visit		9.21 (1.05)		10.04 (4.94)	
	F								
Global Thoracic	M	0.99 *s(p=.001)	-1.58 *s(p=.001)	Collapsed visit		6.65 (0.45)		6.60 (0.45)	
	F								

Tables C10-11 display all main effects of a four-way mixed ANOVA) run on the mean surface EMG for each muscle. The two repeated factors used in the analysis were insole condition (visit 1, visit 2) and side (left-right limb stance phase) and the two between group factors were sex and compliance. Yellow indicates interactions or main effects involving visit and green indicates a difference between left and right limb stance that is only present for one visit.

Tables C10: Mean lower body surface EMG activation

<table>
<tr><td colspan="6">Mean Lower Body EMG (%MVC)</td></tr>
<tr><td rowspan="2">Muscle</td><td rowspan="2">Sex</td><td colspan="2">Visit 1</td><td colspan="2">Visit 2</td></tr>
<tr><td>Left</td><td>Right</td><td>Left</td><td>Right</td></tr>
<tr><td rowspan="2">Tibialis Anterior</td><td>M</td><td colspan="2" rowspan="2">NC 6.32 (1.75)
C 9.05 (0.85)</td><td colspan="2" rowspan="2">NC 7.79 (0.97)
C 6.56 (1.03)</td></tr>
<tr><td>F</td></tr>
<tr><td rowspan="2">Peroneou s Longus</td><td>M</td><td colspan="2" rowspan="2">16.47 (1.29)</td><td colspan="2" rowspan="2">14.71 (1.17)</td></tr>
<tr><td>F</td></tr>
<tr><td rowspan="2">Gastrocn emius</td><td>M</td><td rowspan="2">NC 16.69 (1.43)
C 20.95 (2.61)
*s(p=.022)</td><td rowspan="2">NC 20.10 (2.96)
C 14.77 (3.45)
*s(p=.022)</td><td rowspan="2">NC 17.49 (1.12)
C 18.26 (1.88)</td><td rowspan="2">NC 16.38 (1.42)
C 17.32 (1.89)</td></tr>
<tr><td>F</td></tr>
<tr><td rowspan="2">Bicep Femoris</td><td>M</td><td colspan="4">1.22 (0.30) *sx(p=.020)</td></tr>
<tr><td>F</td><td colspan="4">2.99 (0.88) *sx(p=.020)</td></tr>
<tr><td rowspan="2">Vastus Medialis</td><td>M</td><td colspan="2" rowspan="2">5.17 (0.82) *v(p=.034)</td><td colspan="2" rowspan="2">6.16 (0.97) *v(p=.034)</td></tr>
<tr><td>F</td></tr>
<tr><td rowspan="2">Gluteus Minimus</td><td>M</td><td colspan="2" rowspan="2">4.09 (0.50)</td><td colspan="2" rowspan="2">2.93 (1.17)</td></tr>
<tr><td>F</td></tr>
</table>

Tables C11: Mean upper body surface EMG activation

Mean Upper Body EMG (%MVC)					
Muscle	Sex	Visit 1		Visit 2	
		Left	Right	Left	Right
External Obliques	M	3.09 *sx(*p*=.011)			
	F	7.38 *sx(*p*=.011)			
Internal Obliques	M	2.84 (0.75) *sx(*p*=.028)			
	F	7.21 (3.10) *sx(*p*=.028)			
Rectus Abdominus	M	1.72 (2.26)		1.18 (3.58)	
	F				
Latissimus Dorsi	M	1.73 (0.47)		2.52 (0.32)	
	F				
Lower Thoracic Erector spine	M	1.41 (0.44) *sx(*p*=.015)			
	F	2.28 (0.67) *sx(*p*=.015)			
Lumbar Erector Spine	M	NC 2.03 (1.13) C 2.12 (0.48)	NC 1.67 (0.25) C 2.95 (0.61)	NC 2.02 (1.04) *sx(*p*=.016) C 2.92 (0.68)	NC 2.47 (0.43) C 2.51 (0.21)
	F	NC 2.33 (0.68) *v(*p*=.016) C 4.47 (3.40)	NC 3.63 (1.95) C 2.75 (0.51)	NC 5.48 (3.37) *l(*p*=.039) sx(*p*=.016) v(*p*=.016) S(*p*=.040) C 2.46 (0.76) *l(*p*=.039)	NC 3.07 (0.92) *s(*p*=.040) C 3.25 (1.99)

Table C12 displays the supplementary five-way mixed ANOVA, separate from the MANOVA, that was run on torso muscles, with the additional factor looking at both the stance and swing side of the body.

Tables C12: Mean upper body surface EMG activation

Mean Upper Body EMG With Stance and Swing Phase (%MVC)									
Muscle	Sex	Visit 1				Visit 2			
		Left		Right		Left		Right	
		Stance	Swing	Stance	Swing	Stance	Swing	Stance	Swing
External Obliques	M	3.87 (1.43)		4.46 (0.89)		2.32 (0.57) *sx(p=.004)		2.52 (0.69) *sx(p<.001)	
	F	7.76 (1.52)		7.12 (1.77)		6.99 (1.22) *sx(p=.004) st(p<.001)		9.44 (1.92) *sx(p<.001) st(p<.001)	
Internal Obliques	M	Stance 2.80 (1.98) Swing 3.05 (0.32)							
	F	Stance 7.24 (1.98) *st(p=.028) Swing 10.01 (2.00) *st(p.028)							
Rectus Abdominus	M	1.69 (0.47)				1.45 (0.91)			
	F								
Latissimus Dorsi	M	Stance 2.13 (0.53) *st(p=.002) Swing 1.51 (0.47) *st(p=.002)							
	F								
Lower Thoracic Erector spine	M	1.78 (0.38)	1.72 (0.51)	1.92 (0.44) *st (p<.001)	1.36 (0.36) *st (p<.001)	Collapse visit			
	F								
Lumbar Erector Spine	M	NC 2.10 (0.51) C 2.12 (0.13)		NC 1.80 (0.30) C 2.99 (0.72)		NC 1.91 (0.48) * sx(p=.008) C 2.96 (0.47)		NC 2.53 (0.480 C 2.46 (0.27)	
	F	NC 2.54 (0.39) *v(p=.012) C 4.42 (1.57)		NC 4.10 (2.00) C 2.94 (0.51)		NC 5.76 (1.64) *v(p=.012) s(p=.049) l(p=.029) sx(p=.008) C 2.56 (0.36) * l(p=.029)		NC 3.43 (0.87) C 3.32 (2.02) *s(p=.049)	

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