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Kinematic and Kinetic Effects of Alterations in Lumbar Lordosis in People with Tight Hamstrings

Dema Assaf
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Kinematic and Kinetic Effects of Alterations in Lumbar Lordosis in People with Tight Hamstrings

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From The University of Tennessee
and
The University of Memphis

By
Dema Assaf
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ABSTRACT

The etiology of nonspecific low back pain is sparsely understood. To better understand the contributing factors to nonspecific low back pain, there are often common concurrent pathologies that are investigated to determine their functional relationship to low back pain. One such pathology, investigated further in this thesis, is tight hamstrings. Specifically, the effect of hamstring length on pelvic position during gait and activities of daily living under normal and altered spinal position were investigated as part of this study in a motion analysis lab.

First, a marker validation study was conducted to ensure the accuracy of sagittal spinal measures of lumbar lordosis, thoracic kyphosis, and sagittal vertical axis, which are calculated using skin markers. Lateral x-rays taken by the EOS bi-planar scanner were used to measure both clinical and marker measures. Sagittal spinal measures were also output by the built-in sterEOS software. These measures were compared and found to be accurate within clinical requirements, despite inaccuracy of individual marker placement in identifying intended spinal anatomy.

After validating the accuracy of spinal measures of interest for this study, kinematics of the pelvis and spine were analyzed during normal gait under two conditions: at a normal and altered spinal position. This revealed a unique pelvic compensation pattern in those with tight hamstrings to changes in lumbar lordosis. While other study participants exhibited varied pelvic responses to changes in lumbar lordosis, those with tight hamstrings responded with a -0.7° ± 1.6° decrease in pelvic tilt for every 1° of decreased lumbar lordosis (R^2 = 0.94).

Finally, a similar kinematic analysis was conducted during stair ascent and descent. The results of this analysis, however, revealed a more random pelvic compensation pattern to changes in lumbar lordosis even among those with tight hamstrings. A kinematic and kinetic analysis of the angles, moments, and powers at the hip, knee, and ankle during stair ascent and descent also revealed no significant differences between those with and without tight hamstrings, with the exception of hip kinematics during swing phase of stair ascent (p = 0.047).
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<tr>
<td>ASISs</td>
<td>Anterior Superior Iliac Spines</td>
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<td>ASTANCE</td>
<td>Stance Phase during Stair Ascent</td>
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<tr>
<td>ASWING</td>
<td>Swing Phase during Stair Ascent</td>
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<td>DSWING</td>
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<td>LBP</td>
<td>Low Back Pain</td>
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<td>LL</td>
<td>Lumbar Lordosis</td>
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<td>NLBP</td>
<td>Nonspecific Low Back Pain</td>
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<td>PSISs</td>
<td>Posterior Superior Iliac Spines</td>
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<td>PT</td>
<td>Pelvic Tilt</td>
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<td>RMS</td>
<td>Root Mean Squared</td>
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<td>SVA</td>
<td>Sagittal Vertical Axis</td>
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<tr>
<td>TK</td>
<td>Thoracic Kyphosis</td>
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<tr>
<td>TPA</td>
<td>Trunk-pelvis Angle</td>
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<td>UTHSC</td>
<td>University of Tennessee Health Science Center</td>
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CHAPTER 1. INTRODUCTION

Significance of Research/Clinical Problem

According to the 2013 Center for Disease Control health report, 28% of adults in the United States in 2012 reported experiencing back pain in the three months prior to questioning, and these statistics have not significantly changed since the 1990s [1-3]. In fact, low back pain (LBP) is estimated to affect 8 out of 10 people at some point during their lives [3]. In the United States, people suffering from LBP spend $96 million annually on treatments [4]. This estimation, however, does not take into account indirect costs due to lost productivity, which increases this estimation to a staggering $100-200 billion annually [5]. This is important, since second only to the common cold, LBP is the leading reason for work absences, accounting for 40% of absences in the United States [6]. Additionally, LBP is the leading reason for worker’s compensation claims [3, 6].

Low back pain is classified into three categories based on its source. LBP can spur from neurological issues such as radiculopathy or stenosis [2], develop due to spinal anatomical/congenital abnormalities, or be idiopathic in nature where the pain has no known pathological cause. This third category is known as nonspecific LBP (NLBP) [2]. Since its etiology is unknown, its diagnosis and treatment is also elusive and palliative in nature. Even with treatments including pain medications, physical therapy regimens, and as a last resort, surgical interventions, patients suffering from LBP are often never relieved of their symptoms. Many studies show that those experiencing chronic LBP (defined as LBP persisting over the course of a 3-month span) often relapse and experience episodes of pain throughout their lifetimes in a waxing and waning manner [3, 7].

These pain treatment methods and trends showing recurrence of symptoms, have not changed since the early 1990s, though treatment costs associated with LBP have increased exponentially. In 1998, overall health care costs associated with LBP treatments were approximately $26.3 billion, 3.6 times less than the more recent cost estimation [2]. This corresponds to a total cost, which includes indirect expenses, between $50-100 billion [6]. One study analyzed the per capita impact these figures have on patients with spine problems in the United States, finding an increase in annual medical costs (normalized for age and gender) of $6096 in 2005, compared to $4695 in 1997 [8]. In those same years, patients without spine problems had significantly lower medical expenses: $3516 in 2005 and $2731 in 1997 [8].

Due to the elusive nature of the etiology of NLBP and the difficulty in its treatment, researchers are actively searching for various links and markers to help further classify and more effectively treat NLBP. One such link that continues to be investigated is the link between LBP and tight hamstrings. Studies have shown that people with LBP have a higher incidence of tight hamstrings [9, 10], though this is still a subject of debate within some findings [11]. A recent study, however, has shown that the severity of LBP increases as hamstring tightness increases [12]. Tight hamstrings may posteriorly tilt the
pelvis and decrease hip rotation [13]. This posterior pelvic tilt causes the lumbar spine, which normally exhibits a slight concave curve posteriorly called a lordosis, to flatten out becoming more hypolordotic [14], a positioning that has been shown to increase vertebral segment shear forces [15].

For this particular study, two major comparisons are made to evaluate the kinematic and kinetic responses to changes in spinal curvature, as may be seen after surgical spinal fusions performed in those with LBP or acquired from degenerative disc disease. The first comparison analyzes the kinematics of participants before and after an induced change in spinal curvature using a custom orthosis designed to alter sagittal spinal alignment without contacting the pelvis. The second comparison looks at the kinetic and kinematic responses of participants with tight hamstring muscles compared to those with normal hamstring lengths.

This thesis is organized into 3 main chapters, the first of which (chapter 2) is a justification of the methodology and accuracy of measures through a marker validation study. Chapter 3 discusses the spino-pelvic interactions during normal gait under the normal and induced spinal change conditions. Chapter 4 provides an overview of spino-pelvic interactions and kinetic and kinematic responses of the lower extremity joints to changes in spinal position during stair ascent and descent.

**Overview of Anatomy**

**Spine**

The spine consists of five main regions: cervical, thoracic, lumbar, sacral, and coccygeal. It is made up of 33 individual vertebrae. The first 24 from the top (7 cervical, 12 thoracic, and 5 lumbar) can move independently of each other (Figure 1-1). The remaining 9 (5 sacral, and 4 coccygeal) are fused together. The sacral segments do not play a role in spinal motion but are important in transferring weight from the axial to the lower appendicular skeleton and anchoring ligamentous support to the pelvis.

Though shape, size, and orientation differ based on spinal region, vertebrae have the same basic structure throughout the spine. They consist of a flat, bony region called the vertebral body. Posterior to the vertebral body is the spinal canal (in an opening called the foramen), through which the spinal cord passes. The individual spinal canals line up creating an opening from the top to the bottom of the spine. Vertebrae also include spinous processes, which project on the posterior aspect of the spine, and transverse process that project medially and laterally. Facet joints are the posterior articulating surfaces between vertebrae. Each vertebra consists of 4 facet joints: two superiorly to connect with the vertebra above it and two inferiorly to connect with the vertebra below it. Between each vertebral body is a vertebral disc, preventing the bones from rubbing together unfavorably. Vertebral discs consist of a fibrocartilage outer region and a gel-like inner region: the annulus fibrosus and the nucleus pulposus, respectively. The
Figure 1-1: Spine Anatomy
Representation of the 5 regions of the spine (cervical, thoracic, lumbar, sacral, coccyx) and the vertebral segments within each region
Accessed 1-29-2015 [16]
vertebrae are kept together by two ligaments that run the span of the spine at the vertebral bodies: the anterior and posterior longitudinal ligaments.

The first two vertebrae in the cervical spine, which make up the bones of the neck, have slightly different shapes from the remaining 5 cervical vertebrae and feature an increased range of motion [16]. The first of these two vertebrae, called the atlas, is ring-shaped and allows for the up-and-down nodding motion of the head. It is the only vertebra that does not include a vertebral body. The second, called the axis, allows for the side-to-side rotation in the transverse plane of the head [16]. The axis features a dens (odontoid process) that fits into the ring of the atlas, allowing for this specialized motion. The remaining 5 cervical vertebrae are the smallest compared to both the thoracic and lumbar vertebrae. They have three foramen, one vertebral and two transverse, and feature facets that are oriented near the transverse plane, allowing for more rotation.

The thoracic vertebrae are larger than the cervical vertebrae, and feature a single vertebral foramen. Their spinous processes extend more inferiorly and stack one above the other. The facet joints are oriented in the coronal plane. This position facilitates lateral flexion and rotation and limits flexion and extension. Vertebrae in this region feature additional facets which serve as articulating surfaces with respective ribs. The thoracic vertebrae are also responsible for supporting the rib cage that in turn protects thoracic vital organs [16].

The lumbar vertebrae are much larger and more wedge-shaped than the cervical and thoracic vertebrae, allowing for the lordotic curvature [16]. Their vertebral bodies are taller and wider with larger spinous processes that project more posteriorly. Their transverse processes are also larger and blunter. The function of the lumbar spine is mostly weight bearing and plays a large role in lifting [16]. Its facets are found on the sagittal plane, allowing for more flexion and limiting lateral motion as seen in the thoracic spine.

Viewed from the side, the spine has a s-shaped curvature, made up by a slight concave posterior curve formed by each the cervical and lumbar spinal segments (known as lordosis) and a slight concave anterior curve formed by the thoracic spinal segments (known as kyphosis) (Figure 1-1). Lordotic and kyphotic curvature is normal; however, in excessive or lesser curvatures, loading patterns are altered, causing increased vertebral loads [2, 14]. In the lumbar spine, excessive curves cause hyperlordosis while insufficient curvature causes hypolordosis. Either of these states causes increased stresses in the spine and may lead to LBP.

Imbalance of surrounding muscles as well as degenerative and neurologic pathologies causes hypo- and hyperlordosis in the lumbar spine. Specifically, hypolordosis develops when the hip flexors and trunk extensor muscles are weak in conjunction with tight abdominal muscles and hip extensors, including the hamstring muscles [17, 18]. One study showed this muscle imbalance pattern to be more prevalent in those with LBP when compared with a control group [19]. This altered lumbar curvature shifts the loads in the spine anteriorly, increasing stresses through compressive
forces on both the intervertebral discs and the vertebral bodies. Hypolordosis has also been shown to increase intervertebral shear stresses [15]. Hyperlordosis is the opposite phenomenon, occurring with weaker abdominal and hip extensor muscles and strong hip flexor and trunk extensor muscles [17, 18]. Conversely, a hyperlordosis shifts the spinal loads posteriorly, compressing the facets and spinal column.

With a normal lordosis, spinal loads are distributed across the different spinal components evenly, with spinal muscles and tendons ensuring balanced load distribution [20]. When this curvature is altered, an unbalanced load distribution surfaces, increasing the loads acting on certain spinal components. Normally, 80% of spinal compressive forces are resisted by the vertebral bodies and their intervertebral discs [20]. Hypolordosis increases this normal compressive force on the vertebral bodies and discs, throwing off the normal load distribution [20]. The increase in vertebral segment shear forces found in those with hypolordosis also shifts the normal spine loading balance, increasing the resistance from the apophyseal joint surfaces, which shield the intervertebral discs from shear forces and torsion [20]. This imbalance can strain regions of the spine more than what they can resist, potentially causing LBP.

**Pelvis**

The pelvis consists of a right and left hemisphere with ilium, ischium, and pubis bones (Figure 1-2). The sacral segments of the spine connect to the ilia of the pelvis at the sacroiliac joint. The ilium, as seen below in Figure 1-2, spans the top region of both the right and left hemisphere of the pelvis. Inferior to the ilium are the pubis bones, which connect at the pubic symphysis. Inferior and posterior to the pubis bones are the right and left ischia (Figure 1-2). The ilium, ischium, and pubis help to define the hip joint, in particular the orientation and positioning of the acetabulum, a cup-like region in which the femoral head articulates.

The connection between the spine and the pelvis at the sacroiliac joint helps support the spine and pelvis. This connection is supported by ligaments such as the sacrospinous ligaments, which attach to the ischial spine and insert on the lateral regions of the sacrum and coccyx, and sacrotuberous ligament, which is found on the posterior region of the pelvis and attaches to the posterior sacrum. These ligaments help to prevent sacral rotation about the ilium, which occurs due to the large weight endured by the sacrum, allowing slight gliding motions and providing structural stability during bending [21]. Another ligament that plays an important role in spinopelvic stability is the dorsal sacroiliac ligament, which is found to be in tension during posterior pelvic tilt, which occurs during a decrease in lumbar lordosis (LL).

The pelvis serves to balance LL and hip extension in order to provide for erect posture with minimal energy expenditure [22]. The ability of the pelvis to balance this posture is dependent on its shape and its relation to the sacral slope, a parameter used to define the pelvic tilt in the sagittal plane, measured as the angle between the top of the
Figure 1-2: Pelvis Anatomy
Representation of the anatomic regions of the pelvis in both anterior and posterior views
sacrum and the horizontal. Pelvic shape and sacral slope help define the degree of LL present [22]. As mentioned, deviations from the normal degree of LL are seen in people with tight hamstrings, who have been shown to have a more posterior tilt of the pelvis and decreased hip rotation [13], thus a more hypolordotic lumbar curve [14] (Figure 1-3).

This balancing by the pelvis affects the orientation of the acetabulum and thus femoral head coverage. As the pelvis tilts anteriorly, as seen when the hamstrings are stretched, the acetabulum rotates forward, thus increasing femoral head acetabular coverage [24, 25]. Conversely, a decrease in femoral head acetabular coverage is seen when the pelvis tilts posteriorly and the acetabulum rotates backward, as seen in those with tight hamstrings [24, 25]. In motion, decreased acetabular coverage leads to increased pressures at the joint due to decreased contact surface area. This can potentially lead to osteoarthritis of the hip.

Hamstring Muscles

The hamstring muscle group incorporates three muscles (the semitendinosus, semimembranosus, and the bicep femoris) that serve as hip extensors and knee flexors (Figure 1-4). The hamstrings are located on the posterior region of the thigh and originate at the ischial tuberosity of the pelvis, which is located at the inferior-most point of the ischium. The semitendinosus and semimembranosus insert into the medial side of the tibia, and the biceps femoris inserts into the lateral side of the fibula.

Hamstring muscles can become tight due to lack of stretching, particularly in those who remain active with sports, or a sedentary lifestyle. Athletes often use their hamstring muscles in tension during activity, so neglecting to stretch these muscles after exercise can cause them to tighten. It is also thought that sitting for extended periods of time, during which hamstring muscles remain tight, can also lead to increased hamstring tightness, though there is some dissent over this hypothesis [10].

The relationship between tight hamstrings and the spine with respect to kinematics and the cause of back pain is elusive. With this study, we hope to gain a better understanding of the effect tight hamstrings have on lower back kinematics and lower extremity kinetics during normal daily activities.

Overview of Gait Analysis Techniques

Motion analysis is a technology-based technique to study human motion through the use of cameras and skin markers. Motion capture technology is a noninvasive method of investigating kinetics and kinematics of human motion during activities of daily living. Skin markers are used for tracking of segments to allow for better understanding of the functional relationships between different parts of the body. The kinematic and kinetic relationships between the hamstrings, pelvis, and spine during normal daily activities,
Figure 1-3: Hamstring Effects on Pelvic Tilt
Representation of pelvic tilt in stance normally (left), with tight hamstrings (center), and with lengthened hamstrings (right)

Figure 1-4: Hamstring Anatomy
Posterior View of the three hamstring muscles in stance
such as walking and stair ascent/descent, were assessed using the motion analysis technology found in the University of Tennessee Health Science Center (UTHSC) Biomechanics Laboratory. The EOS bi-planar x-ray scanner at LeBonheur Children’s Hospital (Memphis, TN) was used to validate the accuracy of measurements made using the adopted skin marker set. All technology used as part of this study is outlined in detail below.

**Qualisys Motion Capture System**

The Motion Analysis Lab features 10 opto-electronic cameras (Qualisy’s, AB, Gothenburg, Sweden) to provide a comprehensive view of the testing platform, as described in previous studies [27]. Three camera models (100, 300, and 310) are used in the lab setup and were only used to record motion in the infrared spectrum. The different models feature different resolution specifications, the 100 series cameras exhibiting the lowest resolution (0.3 megapixels). Overall, however, for this study, the system was calibrated to track marker position within a resolution of 0.5 mm, sufficient for the measures calculated. The cameras were used to track retro-reflective markers (12.7 mm in diameter) used to define spinal, pelvic, and lower extremity motion.

**AMTI Force Plates**

The lab features three Model 0R6-7 2000 series force plates (AMTI, Watertown, MA, USA), embedded in the testing platform, as used in previous published studies [27, 28]. Ground reaction forces detected by these force plates were used to determine joint reaction forces at the ankles, knees, and hips, the positions of which were determined based on the kinematics of segments as defined by skin marker positioning. For example, at the knee and ankle, the joint center was defined as the midpoint between markers defining the medial and lateral femoral epicondyles and malleoli, respectively. The force plates were also used to indicate the start and end of cycles in the stair ascent and descent testing protocols outlined in further detail in Chapter 4. Running on a Wheatstone bridge principle, the force plates use strain gauges to measure forces and moments in three dimensions [29]. The root-mean-square (RMS) instrumentation error for force components measured by this setup are ± 13 N (x = y = 4448 N) and ± 25 N (z = 8896 N). Hysteresis accounts for ± 0.2% (full scale output) of the system error in the x, y, and z directions [29]. A non-linear error of ± 0.2% (full scale output) also affects the system [29].

**AMTI Force Platform Stairway**

The AMTI Force Platform Stairway (AMTI, Watertown, MA, USA) was designed so that force analysis can also be conducted during stair ascent and descent by transmitting forces directly from the steps to the force plates. A simple set of three matrices, developed through basic system calibrations, allows for the translation of forces
measured by the force plates to their actual position on the stairs. The data analysis software Visual3D (C-Motion, Inc.) treats each step as a force platform onto which forces are transferred based on the translation described by these matrices. The platform consists of 3 steps (rise height 7”, tread length 10.4”, width 23.8”) and is mounted directly into two of the three available force plates. Steps 1 and 3 attach to one force plate while step 2 attaches to another force plate. This alteration of force plate connections allows for easier distinction between which step records contact in data analysis.

EOS Bi-planar X-ray System

The EOS Bi-planar x-ray system (EOS Imaging, Paris, France) used as part of this study resides at Le Bonheur Children’s Hospital in Memphis, TN. Released in 2007 to the market, the EOS Bi-planar x-ray system features two x-rays that capture images with a 1:1 scale simultaneously, one laterally and one frontally. The quality of these images is comparable to those of computed and digital radiography [30]. The images from the EOS system boast pixel sizes as small as 254 µm [31] and 30,000-50,000 shades of gray (compared to hundreds, as seen in standard x-ray images) [32]. This allows for higher image resolution and better contrast within the image. Radiation from imaging techniques such as the EOS scanner, which uses slot-scanning technology, is one-tenth that from digital radiography [31]. When compared to CT scans, the radiation dose decreases to 1/1000 [31].

These two x-ray images can then later be used in conjunction with a database of CT reconstructions to develop a 3D model through the company’s sterEOS software, developed by the Biomechanical Laboratory of the Arts et Métiers ParisTech (Paris, France) and the Orthopaedic and Imaging Laboratory of the École de Technologic Supérieure de l’Université du Québec (Montreal, Canada) [32]. From the two images, trained technicians identify and mark certain bony landmarks from which the system then identifies the position, rotation, and shape of the segment to be estimated in 3D (Figure 1-5).

From these 3D models, certain parameters, including parameters defining spinal curvature and the pelvis such as pelvic tilt, lumbar lordosis, and thoracic kyphosis, can be determined. In order to view both acetabulum in the lateral image and thus develop a 3D model of the pelvis, patients are positioned with one foot slightly offset anteriorly from the other. The sterEOS software takes into account any rotation in the development of the 3D model and accounts for this shift in the determination of clinical parameters.

For this study, the EOS scanner was used to scan test participants with skin markers in place. The lateral images were used to identify and ensure accuracy of marker placement in representing anatomical references. The images were also used to ensure accuracy of changes in marker-derived measures of the pelvis and spine when compared to the radiographically-derived measures determined from the anatomy.
Figure 1-5: Development of the sterEOS 3D Spine Model
(Left) Lateral and frontal EOS images with marked vertebrae as needed for the development of the sterEOS 3D spine model (Right). These images are output as part of a report generated by the sterEOS software.
CHAPTER 2. VALIDATING THE USE OF SKIN MARKERS TO DEFINE SAGITTAL SPINAL PARAMETERS

Background

Motion analysis technology can be used to conduct kinematic analyses on the spine with respect to changes that affect normal motion and can lead to pathologies such as low back pain (LPB). Investigating changes in spinal angles during activities of daily living and contrasting them with the kinematics seen in other parts of the body can help develop a better understanding of how these interactions affect the spine and may cause LBP. However, particularly in the spine, small changes in angle measurements exhibit large clinical differences, so ensuring the accuracy of these clinical measures made with the skin markers must be analyzed before motion analysis can be conducted.

Motion analysis is commonly used to better understand human motion through the use of skin markers as a noninvasive method of tracking segment motion through various activities of daily living. Though skin markers are commonly used in motion capture labs, they are limited in their portrayal of what occurs in the actual anatomy by the effects of skin motion and in their placement by palpation for bony landmarks. These limitations are important considerations for measurement accuracy using skin markers, particularly for the spine.

While skin marker placement accuracy has been evaluated in the lumbar spine [33], no standard skin marker set exists to represent the spine. Additionally, no marker set has been validated against radiographic measures for lumbar lordosis (LL) and thoracic kyphosis (TK). The purpose of this study is to begin to explore the association of the surface spinal marker set described by Fowler et al [34] and Syczewska et al [35] with anatomical position identified by spinal x-ray to determine if changes in marker-derived measures of spinal position correlate with radiographic measures.

Two hypotheses were tested to ensure accuracy in marker placement and the sagittal spinal parameters measured using these markers. The first hypothesis was that our skin marker set accurately represents the intended underlying anatomy. This was evaluated through the determination in lateral x-ray images of the closest spinous process to each marker. The second hypothesis tested was that LL and TK determined using this marker set are accurate in representing the parameters as measured using the actual anatomy. LL and TK were evaluated three different ways for this comparison: using the marker sets (marker measures), using the actual anatomy as seen in the lateral x-ray image (clinical measures), and using the sterEOS model as created using two orthogonal x-ray images (EOS measures).
Methods

Participants

This study uses participants recruited for a motion analysis study that explores spinal kinetics and kinematics of healthy subjects with no history of spine, shoulder or lower limb injuries or defects. Twenty subjects (11 male, 9 female) aged 21-35 years (23.6 ± 3.3 years) were enrolled in this study. These participants had a mean mass and height of 71.1 ± 13.8 kg and 1.73 ± 0.067 m, respectively. Participants were recruited using flyers posted at the University of Tennessee Health Science Center (Memphis, TN) and the University of Memphis (Memphis, TN), so most participants were students. With this in mind and a small sample size, other demographic information was excluded from this study.

IRB approval (Appendix A) and informed consent was obtained prior to any testing. A physical therapy examination was conducted by one physical therapist with 25 years’ experience to screen for any unknown pathologies. The physical therapist then placed 8 skin markers at the C7, T4, T7, T10, T12, L2, L4, S2 spinous processes. Additional smaller markers were placed, as needed, between these skin markers to help better define the spinal curve.

Image Capture

With skin markers in place, participants were then scanned using the bilateral EOS x-ray scanner (EOS Imaging, Paris, France) at Le Bonheur Children’s Hospital (Memphis, TN). Each participant was scanned twice, once with and once without the custom orthosis designed to alter the lumbar curvature without contacting the pelvis. This orthosis was used to alter normal spinal position and changes in parameters. As mentioned in Chapter 1, a 3D sterEOS model can be developed from the two orthogonal x-ray images captured by the EOS scanner. For 3 study participants, this sterEOS model could not be developed due to physical limitations (vertebral sacralization or lumbarization). The only other physical abnormality discovered in the radiographic images was a natural fusion in the cervical spine of one participant, though it was above the region of interest for our marker placement (above C7).

The marker measures could not be determined for 3 of the study participants because necessary markers were cut off during the imaging process. Though our study involved healthy participants with average BMI of 23.5 ± 3.3, some subjects had higher BMIs (about 29). While their spinal anatomy can be seen in the images, because of how narrow the scanner is, some skin markers were cut off in the process of image capture. The remaining participants (19 with clinical measures, 17 with marker measures, and 16 with EOS measures) were then compared for accuracy.
Image Analysis

Two spine parameters were assessed in this study: lumbar lordosis (LL) and thoracic kyphosis (TK) (Figure 2-1). Three analyses were run to validate the use of skin markers to define changes in these parameters with and without the orthosis. The first analysis compared changes in spinal parameters determined using the actual anatomy in the lateral images (clinical measures) with the spinal parameters determined using the skin markers (marker measures). The second analysis compared spinal parameters output by the sterEOS software (EOS measures) with the clinical measures. The third analysis compared the EOS measures with the marker measures. A separate analysis determined the accuracy of marker placement in representing the intended spinous processes.

Absolute values of the measured parameters were not compared between measurement modes due to high variation between the findings. This is partially due to the slight differences in measure definitions based on availability of skin markers, which could not be used to define every spinous process. Instead, changes in these parameters, defined as the difference in parameters between images with and without the orthosis, were compared as part of this study. Though absolute values of the parameters could not be compared as part of this study, comparing changes in these parameters was deemed sufficient as part of this validation study.

Clinical Measures

The raw images in Digital Imaging and Communications in Medicine (DICOM) form were exported from the sterEOS system and an orthopedic resident measured all the clinical measures using Surgimap (Nemaris Inc, New York) (Figure 2-2). LL was measured using the L1/L5 Cobb angle, defined as the angle between the superior endplate of L1 and the inferior endplate of L5. TK was measured using the T1/T12 Cobb angle, defined as the angle between the superior endplate of T1 and the inferior endplate of T12.

Marker Position Analysis

In addition to comparisons between the three parameter determination methods, an analysis was conducted to determine the actual placement of the skin markers in relation to the underlying anatomy. The same orthopedic resident used the lateral x-ray images to determine the spinous process closest to each skin marker in Surgimap (Figure 2-2).

Results

Average differences between measurement modes in changes in LL and TK due to the orthosis were determined for comparison. An average difference in changes in LL and TK due to the orthosis between clinical and marker measures was determined to be
Figure 2-1:  Marker Lumbar Lordosis and Thoracic Kyphosis Measures
Source: Reprinted with permission. Hebert, C., *Determination of the functional relationship between lumbar lordosis and pelvic tilt*, in *Orthopaedic Surgery and Biomedical Engineering*. 2014, University of Tennessee Health Science Center: Memphis, TN. [27]

Figure 2-2:  Lateral X-ray Images
Measurement of clinical measures of lumbar lordosis and thoracic kyphosis and marker position (left) and marker measures (right)
(MEAN ± SD) 0 ± 4° and 0 ± 7°, respectively. A similar comparison of changes in LL and TK between the EOS and marker measures revealed an average difference in LL of 3 ± 4° and a difference in TK of 1 ± 4°. Comparing changes in EOS and clinical parameters, a difference in LL of 1 ± 4° and a difference in TK of 2 ± 8° were found. An outline of these findings can be seen in Table 2-1.

Accuracy of marker placement in representing intended underlying anatomy was found to be (MEAN ± SD) 36 ± 12%. Table 2-2 outlines marker placement accuracy for each subject. Highlighted cells indicate incorrectly placed markers. The marker with the highest placement accuracy was S2, with 55% accuracy. Markers representing T4 and L2 were determined to have the least accuracy, at 20% (Table 2-3).

**Discussion**

Though no other study validates skin marker position and spinal measure accuracy in lateral x-ray images, Hashemirad et al conducted a similar study using frontal fluoroscopy images [33]. They also looked at two spinal positions in their study, though they asked their participants to laterally bend their spine to examine motion in the frontal plane. They analyzed deviation from the center of the intended spinal segment both horizontally and vertically in the image and found the marker placement to be accurate within 5.44 cm and 0.72 cm, respectively, for L2, L3, and L4 vertebrae. With this inaccuracy in marker placement, they showed that intervertebral angles at L2-L3 and L3-L4 were accurate within 5° using skin markers, which is similar to this study’s findings in the sagittal plane [33].

Though the individual skin markers do not accurately represent their intended underlying anatomy, changes in spinal parameters calculated using these skin markers are accurate within 4° and 7° when compared with EOS and clinical measures, respectively. This inaccuracy may be due to participant positioning with one foot slightly further in front of the other to allow for views of both acetabulum for pelvic measures. This rotation was not taken into account in any of the methods of parameter measurement, so while it may affect the actual values of the parameter, it should not affect inter-mode variability or measures of change between the two different patient positions. Another limitation of this study is the small variation in BMI (23.5 ± 3.3) in the subject group. Typically, skin markers are less accurate in representing the intended anatomy in those with higher BMIs. A third limitation is that it does not compare absolute values of the LL and TK but differences in these parameters. While this validates the changes in measures as presented in the remaining sections of this thesis, a more accurate method of representing spinal parameters with skin markers is needed.

**Conclusion**

This study shows that, despite the inaccuracy of skin marker placement, parameters measured using the markers together well represent corresponding clinical
### Table 2-1: Average Differences between Spinal Parameter Measurement Modes

<table>
<thead>
<tr>
<th>Mode Comparison</th>
<th>Lumbar Lordosis [deg]</th>
<th>Thoracic Kyphosis [deg]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Marker vs Clinical</td>
<td>0 ± 4</td>
<td>0 ± 7</td>
</tr>
<tr>
<td>EOS vs Marker</td>
<td>3 ± 4</td>
<td>1 ± 4</td>
</tr>
<tr>
<td>EOS vs Clinical</td>
<td>1 ± 4</td>
<td>2 ± 8</td>
</tr>
</tbody>
</table>

### Table 2-2: Accuracy of Marker Identification for Each Participant

<table>
<thead>
<tr>
<th>Subject</th>
<th>C7</th>
<th>T4</th>
<th>T7</th>
<th>T10</th>
<th>T12</th>
<th>L2</th>
<th>L4</th>
<th>S2</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>C7</td>
<td>T5</td>
<td>T8</td>
<td>T10</td>
<td>T11</td>
<td>L3</td>
<td>L4</td>
<td>S1</td>
</tr>
<tr>
<td>2</td>
<td>T1</td>
<td>T6</td>
<td>T9</td>
<td>T12</td>
<td>L2</td>
<td>L4</td>
<td>L5</td>
<td>S2</td>
</tr>
<tr>
<td>3</td>
<td>C7</td>
<td>T5</td>
<td>T8</td>
<td>T11</td>
<td>L1</td>
<td>L3</td>
<td>S1</td>
<td>S2</td>
</tr>
<tr>
<td>4</td>
<td>T1</td>
<td>T7</td>
<td>T10</td>
<td>T12</td>
<td>L1</td>
<td>L3</td>
<td>L5</td>
<td>S2</td>
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<td>C7</td>
<td>T3</td>
<td>T7</td>
<td>T9</td>
<td>T11</td>
<td>L3</td>
<td>L5</td>
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<td>T12</td>
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<td>L3</td>
<td>L5</td>
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<td>T9</td>
<td>T12</td>
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<td>L5</td>
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</tbody>
</table>

*Note: Highlighted cells indicate incorrect marker placement.*

### Table 2-3: Overall Accuracy of Marker Placement

<table>
<thead>
<tr>
<th>Accuracy</th>
<th>C7</th>
<th>T4</th>
<th>T7</th>
<th>T10</th>
<th>T12</th>
<th>L2</th>
<th>L4</th>
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<tbody>
<tr>
<td>Accurate</td>
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<td>8</td>
<td>6</td>
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<td>11</td>
</tr>
<tr>
<td>Within 1 vertebral body</td>
<td>11</td>
<td>8</td>
<td>8</td>
<td>7</td>
<td>10</td>
<td>13</td>
<td>14</td>
<td>9</td>
</tr>
<tr>
<td>Within 2 vertebral bodies</td>
<td>3</td>
<td>8</td>
<td>7</td>
<td>5</td>
<td>4</td>
<td>3</td>
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</tbody>
</table>
measures, thus validating use of this marker set to track spinal motion in motion analysis studies. These findings validate use of motion capture and spinal measures made using skin markers to make clinical observations and assess spinal interactions. As part of this study, we used this skin marker set to better understand the effect of changes in lumbar lordosis on pelvic and lower extremity kinematics and kinetics. This validation study establishes the benefit of use of this marker set in these kinematic and kinetic analyses by quantifying the errors seen in measurements made with the skin markers.
CHAPTER 3. THE EFFECT OF ALTERING LUMBAR LORDOSIS ON PELVIC TILT DURING GAIT: THE ROLE OF THE HAMSTRINGS

Background

Hamstring tightness has been shown to cause abnormal walking patterns including decreased hip and increased knee range of motion, decreased step and stride length, and decreased walking speed [13, 36]. This may be a direct result of a posterior tilt of the pelvis seen in those with hamstring tightness [13]. This positioning puts the lumbar spine in a decreased lordotic state or hypolordosis and increases vertebral segment shear forces and compressive forces on the lumbar vertebrae, which predisposes individuals to disc-related pathologies such as disc herniation or degeneration [15]. Though the connection is still being investigated, this effect of tight hamstrings on the spine can lead to non-specific low back pain (NLBP). Motion analysis was used to better understand these effects and the interactions between the hamstrings, pelvis, and spine during gait.

Frontal-plane pelvic tilt and transverse-plane rotation serve as two determinants of gait. Frontal-plane pelvic tilt is seen as the pelvis tilts downward toward the side of the leg in swing phase and upward toward the side of the leg in stance phase [37-40]. The maximum frontal-plane pelvic tilt can be seen at the point at which the center of mass is most vertical, which coincides with the point at which the swing phase leg passes the stance phase leg [37, 38]. The magnitude is dictated by the hip abductor muscle strength on the side of the leg in stance phase [39], though an average maximal tilt is 4-5° [40]. The other pelvic gait determinant is pelvic rotation, which can be seen in the transverse plane. Pelvic rotation occurs when the pelvis rotates forward on the side of the foot in swing phase as the foot extends forward [37-40]. This serves to increase the stride length [37, 39, 40]. Unlike frontal-plane pelvic tilt, pelvic rotation reaches its maximum value just before foot contact, with a maximal rotation between 3-5° [40]. In order to capture the combined effect of these pelvic motions, this study focuses on the pelvic tilt as seen in the sagittal plane during normal gait. This parameter quantifies the tilt of the pelvis anteriorly or posteriorly in the sagittal plane and is affected by both the frontal-plane pelvic tilt and pelvic rotation.

While many studies have been conducted on the relationship between tight hamstrings and NLBP [9-12], little is understood of the effect of tight hamstrings in gait kinematics, particularly of spino-pelvic interactions. This study uses motion capture technology to assess the relationship between spinal and pelvic interactions in those with and without tight hamstrings. To ensure intra-subject accuracy, changes in spinal and pelvic parameters due to the lordosis-altering effects of a custom orthosis are taken into account as opposed to absolute measures. The hypothesis tested was that those with tight hamstrings, in order to compensate for their more posteriorly-tilted pelvic position, will alter their trunk and pelvic position differently than those without tight hamstrings in response to imposed changes in sagittal spinal alignment.
Methods

Participants

Twenty subjects (11 male, 9 female) were enrolled in this study. These subjects ranged in age between 21-35 years, with a mean ±SD of 23.6 years ± 3.3 years. They weighed an average of 71.1 kg ± 13.8 kg and had a mean height of 1.73 m ± .0670 m. These values corresponded to an average BMI of 23.5 ± 3.3. All subjects were recruited by flyers posted around the University of Tennessee Health Science Center (Memphis, TN) and the University of Memphis (Memphis, TN), so most were students of these institutions. Because of this and the small number of participants, other demographic information was not taken into consideration for this study.

All subjects were considered healthy with no history of spine, shoulder or lower limb injuries or defects. Subjects, after providing IRB-approved informed consent (Appendix A), received a full physical therapy examination by a licensed physical therapist with 25 years’ experience to screen for any undiagnosed physical limitations. Hamstring length was assessed using the passive-knee-extension test [41]. Subjects were positioned in the supine position and lower limb passively placed at 90° hip flexion. Each knee was independently passively extended to reported discomfort and the angle between the shank and vertical recorded. An angle greater than 25° was defined as “tight”. This criterion was met by 6 of the subjects.

Data Collection

Thirty-two 12.7-mm diameter reflective markers were applied over anatomical landmarks to define motion of the torso, pelvis, and lower extremities. The torso was defined using the skin marker set validated in Chapter 2. The pelvis was defined by markers at the posterior- and anterior-superior iliac spines (PSISs and ASISs) and the crests. The lower extremities were defined with markers at the lateral and medial epicondyles, the medial and lateral malleoli, the calcaneus, dorsum, and the fifth, first, and great toes, as defined in a previously-published thesis [27]. Ten optoelectronic cameras (Qualisys, Gothenburg, Sweden) and 3 force plates (AMTI, Watertown, MA) were then used to track segment motion and define gait parameters. The cameras, with a calibrated accuracy of 0.4 ± 0.1 mm, recorded motion at 100 Hz, with movement interpolated over ten frames with a third order polynomial and low-pass filtered at 7 Hz with a fourth-order digital Butterworth filter. Ground reaction forces were filtered using a low-pass filter at 15 Hz.

Subjects were instructed to walk at a self-selected pace across the platform with imbedded force plates until a minimum of five clean foot strikes were seen on the force plates. Clean foot strikes consisted of contact of the heel and ball of the foot on one force plate. This allowed for the averaging of measured parameters across multiple trials for each subject in order to account for variability between trials. Subjects completed this
walking task twice, once with and once without a custom orthosis designed to alter lumbar lordosis without contacting the pelvis.

Data Analysis

A kinematic analysis was run for all subjects during the full gait cycle to assess changes due to the orthosis in lumbar lordosis (LL) (defined as the acute angle between lines connecting T12 to L2 and S2 to L4 markers), pelvic tilt (defined as the acute angle between the horizontal line extended from the posterior superior iliac spines (PSISs) and the line connecting the anterior superior iliac spines (ASISs) to the PSISs), and trunk-pelvis angle (defined as the angle between the line connecting C7 to the PSISs and the PSISs to the ASISs) (Figure 3-1). A full gait cycle was defined to start with foot contact and end with the next ipsilateral foot contact. All parameters determined during this full gait cycle were normalized from 0-100% of the full gait cycle.

In order to quantify the effect of the orthosis on the lumbar spine, LL was calculated for trials conducted both with and without the orthosis. LL was then averaged across the full gait cycle. The average LL determined for all trials conducted without the orthosis were averaged to find the average LL during gait normally. This value was then subtracted from the average LL across the full gait cycle for each trial conducted with the orthosis. The three trials with the greatest change in LL due to the orthosis were then taken into consideration for the remainder of the analysis. This allowed for the analysis of the effect of altered LL on pelvic tilt and trunk-pelvis angle.

For the three trials with the greatest change in LL due to the orthosis, changes in pelvic tilt and trunk-pelvis angle were also determined similarly to LL. For each subject, pelvic tilt and trunk-pelvis angle were averaged across the full gait cycle and then across all trials conducted without the orthosis. These averages were then subtracted from the three trials in which the orthosis resulted in the greatest change in LL. The changes in LL, pelvic tilt, and trunk-pelvis angle due to the orthosis in the three trials with the greatest change in LL for each subject were then averaged.

Results

Seventeen subjects enrolled in this study responded to the orthosis with a decrease in LL, thus a hypolordosis. With only 3 participants responding with an increase in LL, only participants who responded with a decrease in LL were compared as part of this study for consistency. No participant with tight hamstrings responded to the orthosis with an increase in LL. Table 3-1 outlines, for each subject, the average changes in each calculated parameter due to the orthosis during a full gait cycle. Subjects outlined in gray are those who have tight hamstrings based on the study criteria (popliteal angle > 25°). The orthosis decreased LL an average ±SD of -4.39° ± 2.57°. This corresponded to a change in pelvic tilt of 0.57° ± 1.08° and a change in trunk-pelvis angle of -0.83° ± 1.05°.
Figure 3-1: Measurement of Pelvic Tilt and Trunk-pelvic Angle
Source: Reprinted with permission. Hebert, C., Determination of the functional relationship between lumbar lordosis and pelvic tilt, in Orthopaedic Surgery and Biomedical Engineering. 2014, University of Tennessee Health Science Center: Memphis, TN. [27]

Table 3-1: Average Parameters for Each Subject during a Full Gait Cycle

<table>
<thead>
<tr>
<th>Subject</th>
<th>Lumbar Lordosis [deg]</th>
<th>Pelvic Tilt [deg]</th>
<th>Trunk-Pelvic Angle [deg]</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>-3.16</td>
<td>-0.12</td>
<td>-2.00</td>
</tr>
<tr>
<td>2</td>
<td>-5.48</td>
<td>2.00</td>
<td>1.32</td>
</tr>
<tr>
<td>3</td>
<td>-4.79</td>
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<td>-2.08</td>
</tr>
<tr>
<td>4</td>
<td>-3.62</td>
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</tr>
<tr>
<td>5</td>
<td>-6.56</td>
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<td>-0.12</td>
</tr>
<tr>
<td>6</td>
<td>-1.56</td>
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<td>-1.80</td>
</tr>
<tr>
<td>7</td>
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<td>8</td>
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<td>-11.82</td>
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<tr>
<td>Mean</td>
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<td>-0.83</td>
</tr>
<tr>
<td>STD</td>
<td>2.65</td>
<td>1.12</td>
<td>1.09</td>
</tr>
</tbody>
</table>

Note: Highlighted rows indicate subjects with tight hamstrings.
One parameter in particular showed a distinct difference between subjects with tight hamstrings and subjects with hamstring lengths within the normal range. The effect a change in LL has on pelvic tilt can be seen in Figure 3-2. Though all other subjects exhibit varied pelvic compensation patterns in response to lumbar curve alteration, subjects with tight hamstrings have a distinct pelvic compensation pattern. This group compensates with a \(-0.7^\circ (\pm 1.6^\circ)\) decrease in pelvic tilt for every \(1^\circ\) of decreased LL \((R^2 = 0.94)\).

**Discussion**

The findings of this study suggest that people with tight hamstrings have a limited ability to compensate with their pelvis for changes in sagittal spinal alignment. This is especially interesting when contrasted with the varying pelvic compensation patterns available to those without tight hamstrings. With further research, this characteristic compensation pattern may be a potential consideration when planning surgical interventions that may impact the spine or hip, such as total hip arthroplasties and spinal fusions.

This limited compensation pattern in those with tight hamstrings was also seen in a study conducted by Whitehead et al. Whitehead et al simulated hamstring tightness at various lengths using an external brace and found more posterior tilting of the pelvis at the shorter hamstring lengths during gait [36]. In fact, at shorter hamstring lengths, the group found a smaller range of posterior pelvic tilt during the gait cycle [36], implying lesser ability to tilt the pelvis. This coincides with our findings of decreased compensation ability since range of motion at the pelvis during gait is already limited due to the shorter hamstrings.

Congdon et al found that hamstring length is an important factor for increased pelvic rotation when both knee and hip flexion are taken into account in tandem, particularly when the hip is flexed and knee is more extended [13]. However, this study positioned their patients in the supine position to assess these relationships, neglecting the effects of weight in normal gait [13]. While these lower extremity positions are extreme for daily activities such as walking, the findings indicate that changes in pelvic rotation are directly related to collaborative changes in knee and hip flexion due to the hamstrings, as seen in this study’s findings.

**Conclusion**

In conclusion, the hypothesis that those with tight hamstrings will alter their trunk and pelvic position differently than those without tight hamstrings in response to imposed changes in sagittal spine alignment is rejected except with respect to pelvic tilt. Those with tight hamstrings exhibited a limited ability to compensate with their pelvis due to changes in LL. With the limited ability of those with tight hamstrings to compensate with their pelvis for alterations in sagittal alignment, tight hamstrings may be an important
Figure 3-2: Effect of Change in Lumbar Lordosis on Change in Pelvic Tilt in Gait
This graph shows the average change in lumbar lordosis and pelvic tilt across the full gait cycle averaged for each subject across the three trials in which the greatest decrease in lumbar lordosis is seen. The blue represents the pelvic response to changes in lumbar lordosis in subjects with normal hamstring lengths, and the red represents this pelvic response in those with tight hamstrings. Those with tight hamstrings exhibit a limited pelvic compensation to changes in lumbar lordosis.
consideration for therapeutic interventions that affect the spine or hip, though further research is needed to better establish this relationship.
CHAPTER 4. ANALYZING KINETICS AND KINEMATICS DURING STAIR ASCENT AND DESCENT

Background

Stair climbing, like walking, is another common activity of daily living, one many Americans face at work, at home, or at school. Though the number of stairs climbed daily varies based on a number of factors including geography, location of work, and type of home, stairs are a commonality in American life. As seen in Chapter 3, tight hamstrings play a role in pelvic compensation mechanisms to changes in sagittal spinal alignment during gait. Investigating the same spino-pelvic responses due to alterations in sagittal spinal alignment in stairs, which requires larger muscle involvement, greater joint ranges of motion, and higher joint loads, may reveal a deeper understanding of the role tight hamstrings play in motion kinematics.

In the sagittal plane, particularly at the ankle and knee, larger flexion angles are seen in swing phase compared to normal gait. In fact, peak knee moments during stair ascent and descent are double those seen in gait. In stair ascent and descent, power is almost exclusively positive, particularly at the ankle and knee, where both knee extensors and flexors were seen to aid in energy generation. Because of the higher muscle involvement in stair ascent and descent, analysis of the kinetic and kinematic interactions in lower extremity joints in those with tight hamstrings compared to those without tight hamstrings may reveal some variations in response to alterations in sagittal spinal alignment.

While research into the effect of hamstring tightness on normal gait has been investigated by others and ourselves, no such analysis has been conducted during stair ascent and descent. Stair analysis in a gait lab setting has become very popular in current research labs as it shows important considerations and differences when compared to gait and other daily activities. However, even with the increase of investigation of the relationship between tight hamstrings and LBP, no study has been conducted investigating this relationship during stairs ascent and descent. As a continuation to a previous gait study described earlier, in which those with tight hamstrings were found to have a distinct pelvic compensation pattern to alterations in sagittal spinal alignment due to a custom orthosis, a similar analysis using the same orthosis was conducted to spot similar trends in stair ascent and descent.

Also, lower limb joint angles, moments, and powers were analyzed as part of this study. Studies have been conducted comparing lower limb joint kinetics between stair ascent and normal gait as well as between stair ascent and descent, but no other study analyzes differences in lower limb joint kinetics due to hamstring shortness or changes in sagittal spinal alignment, such as that due to surgical interventions at the hip or spine. This study features healthy participants with tight and normal hamstrings as well as the use of a custom orthosis designed to alter sagittal spinal alignment and compares...
lower limb joint kinematics and kinetics across the entire stair cycle in stance and swing phases between these groups in both stair ascent and descent.

Methods

Participants

Twenty participants (11 male, 9 female) were recruited using flyers posted around the University of Tennessee Health Science Center (Memphis, TN) and the University of Memphis (Memphis, TN). The participants enrolled at this study ranged in age from 21-35 years with an average of 23.6 ± 3.3 years. Average mass was 71.1 ± 13.8 kg and average height 1.73 ± 0.067 m. Their corresponding BMIs were 23.5 ± 3.3. With the small number of participants and pool of participants from university campuses, other demographic information was not taken into consideration for this study.

Participants had no history of spine, shoulder or lower limb injuries or defects. To screen for any undiagnosed physical limitations, a physical therapist of 25 years’ experience conducted a full physical therapy examination after participants provided IRB-approved informed consent (Appendix A). Hamstring length was measured using the passive knee-extension test, which was conducted with participants in the supine position [45]. Each knee was independently passively extended to its end range, and the angle between the shank and vertical was recorded. For this study, an angle greater than 25° was defined as “tight”. This criterion was met by 6 participants.

Data Collection

Thirty two 12.7-mm diameter reflective markers were applied to the skin to define motion of the torso, pelvis, and lower extremities. The marker set used to define spinal kinematics was described and validated in an earlier chapter. Motion of these reflective markers was captured by ten optoelectronic cameras (Qualisys, Gothenburg, Sweden) with a calibrated accuracy of 0.4 ± 0.1 mm. The cameras recorded motion at 100 Hz, with movement interpolated over ten frames with a third-order polynomial and low-pass filtered at 7 Hz with a fourth-order digital Butterworth filter.

Ground reaction forces were captured using 3 force plates (AMTI, Watertown, MA) to allow for determination of cycle start and end times and joint moments and powers. These measured ground reaction forces were filtered using a low-pass filter at 15 Hz and transmitted through an AMTI Force Platform Stairway (AMTI, Watertown, MA) that mounts directly into the force plates. This set of stairs features 3 steps 24 in. wide and 10.4 in. long with no railing. Stair height is 7 in. for each step. A matrix determined and validated as part of a previous study was used to determine the exact location of each stair for force analysis in Visual3D (C-Motion Inc). This creates what is called a force platform in the software at which forces are measured. For distinction between steps
during analysis, the AMTI Force Platform Stairway is designed such that forces on the first and third steps are read by the same force plate and forces on the middle step are recorded by a different force plate.

Participants were asked to walk at a self-selected pace across the room toward the stairway. They then ascended three steps and onto a platform not mounted to the force plates. On the platform, participants turned, stood at the top of the steps briefly, then descended the stairs and walked back across the room to the starting position. For each participant during ascent and descent, three clean runs starting with the right and left foot in both ascent and descent were acquired so that measured parameters could be averaged.

As described in a previous chapter, a custom orthosis designed to alter spinal curvature without contacting the pelvis was used to induce a change in LL. For this study, participants were asked to complete the stair ascent/descent task twice: once with and once without the orthosis until 3 clean runs starting with both the left and right feet were recorded under each condition.

**Data Analysis**

All data was collected using Qualisys and exported to Visual 3D for processing. A 2 Hz capture of each participant in a static position was used to create a rigid-linked model with 6 degrees of freedom composed of 8 geometric shapes [46, 47]. This model was created as part of a previously-published thesis and consists of right elliptical cylinders to represent the pelvis and torso and cones to represent the right and left feet, shanks, and thighs [27].

All parameters calculated as part of this study were determined across a cycle described as the start of foot contact on a step to the end of contact on that stair with that same foot. The start and end of force plate contact on each stair was determined using a threshold in Visual3D, though with fluctuation in body mass between each participant, manual evaluation of each event was necessary to ensure proper labelling of foot contact and liftoff times.

For this study, in ascent, only the middle stair was used for analysis. The bottom stair was ignored because, since participants walked to the stairway, foot shuffling or longer strides were seen in many trials as participants approached the stairs. The top stair before the final platform was also ignored because, in many trials, participants had already begun to turn before completely stepping onto the final platform. In descent, the middle and bottom steps were used for analysis. The top step was ignored since, at the start of descent, many spinal markers, particularly for taller participants, were obscured from view due to the height of the final platform. While in some trials, not all markers are completely visible at first foot contact on the first stair down, by the time participants begin the descent of their second foot toward the step in question, all spinal skin markers are visible to the cameras. Two analyses were run as part of this study: the first analyzes
the spinal and pelvic kinematics while the second analyzes kinetics at the hip, knee, and ankle joints.

**Kinematic Analysis**

Parameters of the kinematic analysis were measured in the sagittal plane (as validated in chapter 2), taking the sagittal plane to include the midpoints of both the anterior superior iliac spines (ASISs) and the posterior superior iliac spines (PSISs). The parameters measured as part of this study are lumbar lordosis (LL) (See Figure 2-1), pelvic tilt (PT), and trunk-pelvis angle (TPA) (See Figure 3-1). Lumbar lordosis (LL) is defined as the acute angle formed by the intersection of lines formed by connecting S2 to L4 and L2 to T12 markers. Pelvic tilt (PT) is defined as the acute angle between the horizontal line extended from the PSISs and the line connecting the PSISs to the ASISs. Finally, TPA is defined as the angle between the line extended from C7 to the midpoint of the PSISs and the line connecting the PSISs to the ASISs.

For each cycle, these parameters were calculated, normalized to 100% of the full stair cycle, and averaged. LL, PT, and TPA measured in all normal trials for a single participant were averaged and used as a baseline. These average normal parameters were then subtracted from corresponding parameters from each orthosis trial to determine deviation from normal due to the orthotic. The 3 trials for each participant with the greatest change in LL due to the orthosis were then used for the remainder of this analysis. Averages of each parameter (LL, PT, and TPA) across these three trials were then averaged and compared using linear regression analyses.

Consistently, 17 participants responded to the orthosis through the stair cycle with a decrease in LL. Three participants responded with an increase in LL, though in ascent and descent, a different set of 3 participants responded with an increase in LL. Of these three participants, 1 was part of the tight hamstring group in ascent, and 1 and 2 were part of the tight hamstring group in descent, on the second and third step, respectively. For consistency in comparison, only trials in which a decrease in LL was seen were used as part of the kinematic analysis described in this study.

**Kinetic Analysis**

Parameters measured as part of the kinetic analysis were joint angles, moments, and powers at each the ankle, knee, and hip. All parameters were measured in the sagittal plane. Joint angles were defined using the segments created in the model described earlier. The ankle joint angle was defined as the angle between the foot and shank segments. The knee angle was defined as the angle between the shank and thigh segments. The hip angle was defined as the angle between the thigh and pelvis segments.

Joint moments calculated in Visual3D follow the right hand rule (anterior rotation, or flexion, is positive) and are normalized using participant mass. For this study,
only the anterior-posterior plane was taken into consideration for the kinetic analysis, thus knee extension, ankle flexion, and hip flexion were considered positive. In a similar manner to the joint angle determination, ankle moments were determined from the foot with respect to the shank. The knee moments were determined from the shank with respect to the thigh, and the hip moments were defined as the thigh with respect to the pelvis. Joint power at each the ankle, knee, and hip was determined by multiplying the joint moment by the angular velocity of the proximal segment to each joint.

All joint angles, moments, and powers were calculated at each time point of the cycle and normalized to 100% of the full stair cycle. Individual steps were no longer taken into consideration as separate entities for this portion of the analysis. Instead, parameters were averaged across each time point of the cycle for three clean trials for each the left and right foot in ascent and descent. Instead of a comparison between right and left feet, a comparison between what occurs at these joints during stance and swing phases was analyzed. Stance phase joint kinematics were calculated for joints in the leg in contact with the step during the stair cycle. During this same cycle, swing phase joint kinematics were calculated for joints in the leg not in contact with the step. Thus four categories for comparison were developed: stance phase during ascent (ASTANCE), swing phase during ascent (ASWING), stance phase during descent (DSTANCE), and swing phase during descent (DSWING).

Differences in joint angles, moments, and powers during each phase (ASTANCE, ASWING, DSTANCE, and DSWING) were determined between trials with and without the orthosis and between the tight and normal hamstring groups. The former of these resulted in a paired test while the latter resulted in an unpaired test. The former analysis was conducted by taking the average angle, moment, or power in each phase across the full stair cycle. The difference between this average in the trial with the orthosis and without was then determined. Then, a Hotelling’s T-square test (p < 0.05) was conducted to check for significant differences between the two groups. The latter analysis was conducted by taking the average angle, moment, or power in each phase across the full stair cycle for both the normal hamstring group and the tight hamstring group. The difference between these two averages was then determined, and a Hotelling’s T-square test (p < 0.05) was conducted to check for significant differences between the tight and normal hamstring groups.

Since joint angles, moments, and powers are not independent of one another, only one test was conducted for each phase, instead of conducting individual significance tests for each parameter. The Hotelling’s T-square test was used for this analysis, thus taking into account not only the contribution of the individual variables but also the joint contribution of the variables together. Since it groups the different variables together, the Hotelling’s T-square test was used as a preliminary analysis to assess for potential differences in each phase. In phases where significant differences were found in the joint angles, moments, and powers between those with and without tight hamstrings, individual t-tests were run to establish true significance and to determine which variable contributes to the significance.
Results

Kinematic Analysis

In ascent, only the middle stair was taken into consideration for the kinematic analysis. With the application of the orthosis, as seen earlier in gait trials, most study participants compensated with a decrease in LL. Only three participants responded to the orthosis with an increase in LL, one of whom has tight hamstrings. Taking into consideration only the parameters of those who responded with a decrease in lordosis for comparison consistency, participants responded on average with a decrease in LL of \(-2.8 \pm 1.6^\circ\). An outline of the average parameters can be seen in Table 4-1. The pelvic response seen in ascent to changes in LL for all subjects is plotted below in Figure 4-1.

In descent, study parameters were analyzed at two steps (the middle and bottom steps). Similarly, only 2 participants responded with an increase in LL at the middle step and 4 participants responded with an increase in LL at the bottom step. Two of these participants responded with an increase in LL consistently on both analyzed steps. An outline of average parameters for participants responding with a decrease in LL on the two steps in descent can be found in Table 4-1. Unlike in gait, no specific pelvic compensation pattern was seen in those with tight hamstrings in either stair ascent (Figure 4-1) or descent (Figure 4-2).

Kinetic Analysis

At the ankle, no differences (p < 0.05) in angles, moments, and powers exist in any of the comparison groups (Table 4-2). Similar findings were seen at the knee (Table 4-3). At the hip, however, differences were seen in a lumped statistical analysis of hip angles, moments, and powers between the normal and tight hamstring groups in ASWING (Table 4-4). When the parameters of this phase were split and analyzed similarly using individual t-tests, no difference was seen. Additionally, no differences were found in any other phase at the hip. Full comparison analyses at each the hip, knee, and ankle for angles, moments, and powers across the stair cycle can be seen in Appendix B (Figures B-1 to B-9).

Discussion

Unlike findings in normal gait presented earlier in this thesis, no correlations between LL and any of the other analyzed parameters were found, in participants with tight hamstrings and without. This indicates that those with tight hamstrings may only exhibit their limited ability to compensate with their pelvis for alterations in lumbar lordosis in normal gait. This was thought to have been influenced by the increased posterior pelvic tilt shorter participants may exhibit in ascent as they extend the next step, but no correlation between height and changes in pelvic tilt were found.
### Table 4-1: Parameter Averages for Each Analyzed Step in Ascent and Descent

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Ascent – middle step</td>
<td>-2.8 ± 1.6</td>
<td>0.7 ± 1.6</td>
<td>-0.2 ± 1.1</td>
</tr>
<tr>
<td>Descent – middle step</td>
<td>-3.5 ± 2.2</td>
<td>0.6 ± 2.0</td>
<td>-0.8 ± 1.3</td>
</tr>
<tr>
<td>Descent – bottom step</td>
<td>-4.0 ± 2.6</td>
<td>0.4 ± 1.7</td>
<td>-1.3 ± 1.2</td>
</tr>
</tbody>
</table>

### Figure 4-1: Effect of Change in LL on Change in PT in Stair Ascent

This graph shows the average change in lumbar lordosis (LL) and pelvic tilt (PT) across the full stair cycle averaged for each subject across the three trials in which the greatest change in LL is seen. The blue represents the pelvic response to changes in lumbar lordosis in subjects with normal hamstring lengths, and the red represents this pelvic response in those with tight hamstrings. All subjects exhibit an individualized pelvic response to changes in lumbar lordosis. No limited pelvic response is seen for those with tight hamstrings as seen in gait ($R^2 = 0.76$).
Figure 4-2: Effect of Change in LL on Change in PT in Stair Descent
These graphs show the average change in lumbar lordosis (LL) and pelvic tilt (PT) across the full stair cycle averaged for each subject across the three trials in which the greatest change in LL is seen at the middle (top) and bottom (bottom) steps in stair descent. The blue represents the pelvic response to changes in lumbar lordosis in subjects with normal hamstring lengths, and the red represents this pelvic response in those with tight hamstrings. All subjects exhibit an individualized pelvic response to changes in lumbar lordosis. No limited pelvic response is seen for those with tight hamstrings as seen in gait on either the bottom step ($R^2 = 0.01$) or the top step ($R^2 = 0.08$) in stair descent.
Table 4-2: Hotelling's T-squared Test for Ankle Angles, Moments, Powers

<table>
<thead>
<tr>
<th>Difference</th>
<th>Measure</th>
<th>ASTANCE</th>
<th>ASWING</th>
<th>DSTANCE</th>
<th>DSWING</th>
</tr>
</thead>
<tbody>
<tr>
<td>Orthotic – Normal Trials</td>
<td>Angle</td>
<td>0.0 ± 0.79</td>
<td>-0.12 ± 1.14</td>
<td>-0.51 ± 0.88</td>
<td>-0.47 ± 0.90</td>
</tr>
<tr>
<td></td>
<td>Moment</td>
<td>0.01 ± 0.04</td>
<td>0.01 ± 0.02</td>
<td>-0.01 ± 0.04</td>
<td>0.01 ± 0.01</td>
</tr>
<tr>
<td></td>
<td>Power</td>
<td>0.01 ± 0.05</td>
<td>0.03 ± 0.03</td>
<td>0.03 ± 0.08</td>
<td>-0.03 ± 0.04</td>
</tr>
<tr>
<td></td>
<td>P-value</td>
<td>0.76</td>
<td>0.10</td>
<td>0.37</td>
<td>0.12</td>
</tr>
<tr>
<td>Tight – Normal Hamstrings</td>
<td>Angle</td>
<td>2.0 ± 1.20</td>
<td>3.5 ± 1.60</td>
<td>0.42 ± 1.70</td>
<td>1.26 ± 1.39</td>
</tr>
<tr>
<td></td>
<td>Moment</td>
<td>0.0 ± 0.07</td>
<td>0.03 ± 0.02</td>
<td>-0.07 ± 0.05</td>
<td>0.0 ± 0.01</td>
</tr>
<tr>
<td></td>
<td>Power</td>
<td>-0.08 ± 0.08</td>
<td>-0.03 ± 0.04</td>
<td>0.24 ± 0.09</td>
<td>0.15 ± 0.06</td>
</tr>
<tr>
<td></td>
<td>P-value</td>
<td>0.81</td>
<td>0.23</td>
<td>0.43</td>
<td>0.31</td>
</tr>
</tbody>
</table>

Note: Values reported as average differences between comparison groups (MEAN ± SD). Significance level p < 0.05. Angles reported in degrees, moments reported in N-m, and powers reported in W.

Table 4-3: Hotelling's T-squared Test for Knee Angles, Moments, Powers

<table>
<thead>
<tr>
<th>Difference</th>
<th>Measure</th>
<th>ASTANCE</th>
<th>ASWING</th>
<th>DSTANCE</th>
<th>DSWING</th>
</tr>
</thead>
<tbody>
<tr>
<td>Orthotic – Normal Trials</td>
<td>Angle</td>
<td>0.26 ± 1.24</td>
<td>-0.12 ± 1.06</td>
<td>-0.73 ± 2.01</td>
<td>0.27 ± 1.11</td>
</tr>
<tr>
<td></td>
<td>Moment</td>
<td>-0.01 ± 0.03</td>
<td>0.0 ± 0.01</td>
<td>-0.01 ± 0.04</td>
<td>-0.01 ± 0.01</td>
</tr>
<tr>
<td></td>
<td>Power</td>
<td>0.02 ± 0.09</td>
<td>0.01 ± 0.03</td>
<td>0.04 ± 0.08</td>
<td>0.07 ± 0.04</td>
</tr>
<tr>
<td></td>
<td>P-value</td>
<td>0.11</td>
<td>0.23</td>
<td>0.54</td>
<td>0.24</td>
</tr>
<tr>
<td>Tight – Normal Hamstrings</td>
<td>Angle</td>
<td>3.36 ± 1.32</td>
<td>2.93 ± 1.52</td>
<td>4.38 ± 2.80</td>
<td>1.83 ± 1.94</td>
</tr>
<tr>
<td></td>
<td>Moment</td>
<td>0.08 ± 0.05</td>
<td>0.0 ± 0.02</td>
<td>0.15 ± 0.05</td>
<td>0.02 ± 0.01</td>
</tr>
<tr>
<td></td>
<td>Power</td>
<td>0.06 ± 0.10</td>
<td>-0.03 ± 0.03</td>
<td>-0.31 ± 0.09</td>
<td>-0.05 ± 0.05</td>
</tr>
<tr>
<td></td>
<td>P-value</td>
<td>0.29</td>
<td>0.52</td>
<td>0.81</td>
<td>0.39</td>
</tr>
</tbody>
</table>

Note: Values reported as average differences between comparison groups (MEAN ± SD). Significance level p < 0.05. Angles reported in degrees, moments reported in N-m, and powers reported in W.
### Table 4-4: Hotelling's T-squared Test for Hip Angles, Moments, Powers

<table>
<thead>
<tr>
<th>Difference</th>
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<th>ASWING</th>
<th>DISTANCE</th>
<th>DSWING</th>
</tr>
</thead>
<tbody>
<tr>
<td>Orthotic – Normal Trials</td>
<td>Angle</td>
<td>0.85 ± 1.75</td>
<td>0.61 ± 1.76</td>
<td>1.21 ± 2.12</td>
<td>1.63 ± 1.88</td>
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<tr>
<td></td>
<td>Moment</td>
<td>0.0 ± 0.02</td>
<td>0.01 ± 0.01</td>
<td>0.0 ± 0.03</td>
<td>0.0 ± 0.01</td>
</tr>
<tr>
<td></td>
<td>Power</td>
<td>0.01 ± 0.07</td>
<td>-0.01 ± 0.02</td>
<td>-0.01 ± 0.02</td>
<td>-0.02 ± 0.02</td>
</tr>
<tr>
<td></td>
<td>P-value</td>
<td>0.38</td>
<td>0.12</td>
<td>0.29</td>
<td>0.38</td>
</tr>
<tr>
<td>Tight – Normal Hamstrings</td>
<td>Angle</td>
<td>2.07 ± 2.58</td>
<td>1.44 ± 2.71</td>
<td>1.11 ± 2.52</td>
<td>-1.37 ± 2.19</td>
</tr>
<tr>
<td></td>
<td>Moment</td>
<td>-0.01 ± 0.04</td>
<td>0.02 ± 0.01</td>
<td>-0.04 ± 0.04</td>
<td>-0.02 ± 0.01</td>
</tr>
<tr>
<td></td>
<td>Power</td>
<td>0.0 ± 0.09</td>
<td>0.02 ± 0.03</td>
<td>0.0 ± 0.02</td>
<td>-0.02 ± 0.03</td>
</tr>
<tr>
<td></td>
<td>P-value</td>
<td>0.57</td>
<td>0.048</td>
<td>0.49</td>
<td>0.59</td>
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</table>

*Note:* Values reported as average differences between comparison groups (MEAN ± SD). Significance level $p < 0.05$. Angles reported in degrees, moments reported in N-m, and powers reported in W.
For the kinetic analysis, no significant differences were found in joint angles, moments and powers at the ankle, knee, or hip between trials with and without the custom orthosis. When comparing between tight and normal hamstring groups, the only potential significance found was in swing phase during stair ascent at the hip (p = 0.0475), though after performing individual t-tests on the variables in this phase, no variable was found to contribute to this finding. Though no significance was found in this study, investigating the lower extremity joint kinetics further in this phase may be beneficial with increased sample size. These findings could be a direct result of increased pressures at the joint due to decreased acetabular coverage, often seen in those who exhibit a more posterior pelvic tilt [24, 25]. While this could be an important consideration for stability in surgical interventions like total hip arthroplasty, further research is needed to ensure significance during this phase.

While no other study looks at the effects of hamstring length and changes in sagittal spinal alignment on lower limb joint kinetics, studies have been conducted on joint kinetics during stair ascent and descent. In 2003, Nadeau et al compared differences in joint angles, moments, and powers between stair ascent and normal gait [43]. Their findings show larger differences in ankle and knee flexion between stairs and gait as well as more energy generation than absorption in stair ascent [43]. The latter of these results correlates well with the findings of this study. Also, while some differences exist in peak magnitude during cycles (particularly in ankle moments, where they report double the peak moment of the peak moment found in this study) the trends and ranges of angles, moments, and powers at each of the joints investigated were similar [43]. One explanation for the difference in peak ankle moment could be due to differences in population. Participants of this particular study investigated stair kinematics in participants over 40 years with a median age of 53 years [43]. This theory is verified in a study conducted by Novak et al in 2011, where differences in peak joint moments between younger (mean of 23.7 years) and older (mean of 67.0 years) participants were found [48].

Protopapadaki et al in 2007 investigated differences in joint angles and moments between stair ascent and stair descent in healthy, young individuals [44]. Their findings showed similar differences to the findings of this study across the stair cycle, with peaks in similar points of the cycle. They found significant differences in joint moments between stair ascent and descent during hip flexion and knee extension [44]. The findings presented by Protopapadaki et al validate our decision to analyze joint kinetics during ascent and descent separately.

One limitation of this study, as pointed out in articles by Nadeau et al and Novak et al, is the analysis of joint kinetics strictly in the sagittal plane [43, 48]. Both studies described in these articles indicate the importance of taking into account kinematics in the frontal plane, as abduction at the hip and knee play a significant role in these kinematics [43, 48]. However, while investigating the joint kinematics at the hip, knee, and ankle in the frontal plane may be an important next step in this study, for the full study in which this analysis plays part, the sagittal plane is taken as the main plane of interest. Significant differences in sagittal interactions between the lumbar spine in the
pelvis were seen in gait, and the custom orthotic used as part of this study was designed to alter sagittal spinal alignment, as is common in spine and hip surgical interventions.

**Conclusion**

Unlike findings in normal gait, no specific trend in pelvic compensation was seen in participants with tight hamstrings due to alterations in lumbar lordosis. Also, no significant differences in ankle, knee, and hip angles, moments or powers were found in either stair ascent or descent during stance or swing phases due to induced changes in sagittal spinal alignment using a custom orthotic. Comparisons of joint kinetics between tight and normal hamstring groups showed significant differences only at the hip during swing phase in stair ascent. These findings indicate that with or without surgical interventions that affect sagittal spinal alignment, no significant kinematic differences are seen, except potentially at the hip in swing phase, due to hamstring tightness. This could be a direct result of decreased acetabular coverage at the hip joint.
CHAPTER 5. GENERAL DISCUSSION

Clinical Significance

The results of this research provide a better understanding of the effects of changes in spinal alignment on kinematics and kinetics during activities of daily living and the differences in these effects seen in a population of people with tight hamstrings. Skin markers defining the spine, though inaccurate in their placement, provide an accurate way of measuring changes in clinical spinal parameters. This helps ensure the accuracy of all measures made as part of this motion analysis study. Using this validated marker set, this study uncovers a limited compensation pattern in the pelvis for changes in sagittal spinal alignment during gait. This is a significant finding and potential consideration for planning surgical techniques that may alter sagittal spinal curvature. Particularly, increased posterior pelvic tilt, seen in those with tight hamstrings, is linked to decreased acetabular coverage at the hip joint [24, 25]. Decreased coverage of the acetabulum at the hip joint decreases joint contact area and thus increases the pressures at the joint. This is potentially an important consideration for the stability of total hip arthroplasty in those with tight hamstrings in order to combat the risk of joint dislocation.

Another potential effect of tight hamstrings is osteoarthritis. Tight hamstrings is often seen in men, particularly those active in sports, during which the hamstring muscles are used mostly in tension. Neglecting to stretch the hamstring muscles can lead to muscle tightening. Sports activity has also been shown to increase the risk of hip osteoarthritis [49]. Hamstring tightness may explain the relationship between sports activity and the higher risk of osteoarthritis. In fact, decreased isokinetic hip extension strength is lower in those with severe hip osteoarthritis, though no differences were seen in hamstring muscle cross-sectional area [50]. More research is needed, however, to better understand this potential link.

Despite kinematic pelvic compensation differences found during the gait cycle for those with tight hamstrings, kinematic and kinetic analyses during stair ascent and descent revealed no such differences. This implies that a unique pelvic compensation pattern to changes in sagittal spinal alignment is a phenomenon specific to gait kinematics. With a slight difference seen only in the swing phase during stair ascent in hip kinetics between those with tight and normal hamstrings, no significant difference in lower limb kinetics and kinematics can be said to exist based on the findings of this study, though conducting this study with more subjects and separating the lumped statistical analysis to better understand which parameters exhibit the greatest effect on significance levels may be beneficial.

Limitations and Future Work

A number of limitations are associated with the studies conducted as part of this research, many of which have been addressed in previous chapters. One such limitation is
that rotation due to subject positioning in the EOS scanner is not taken into account in the measurement of LL and TK. Instead, all measurements are made using only the lateral images, assuming that this represents the sagittal plane. One advantage of the EOS scanner is that it captures two x-ray images simultaneously, one laterally and one frontally. From these two images, the built-in sterEOS software generates a model based on a database of computed tomography reconstructions that accounts for this rotation in the actual anatomy. However, with the proprietary nature of the sterEOS software, there is no current method of accounting for this rotation in measures made using the spherical skin markers. In the works is an extension to this software that would allow for finding the three-dimensional position of the skin markers. This would allow for direct per-subject correlation of the spinal curve defined by the skin markers to the actual spinal curve as defined by the spinous processes, negating the need for comparisons between changes in clinical measures.

Another limitation of this study is the small sample size. With 20 enrolled subjects, only 6 were found to have tight hamstrings. Additionally, in each analysis, three subjects were eliminated for better comparison, as described in each chapter. Promising trends have been found among the tight hamstring subgroup, but expanding the sample size to include more participants with tight hamstrings would allow for stronger correlations and conclusions. Another potential future consideration is investigating how stretching the hamstrings may affect the findings of this research when compared to the normal group.

Another limitation of this study is neglecting hamstring muscle recruitment while analyzing the joint kinetics at the ankle, knee, and hip. While significant conclusions have been made from the kinetic analysis conducted for stairs ascent and descent, no discussion of the role the hamstring muscles play in these findings is present. A future improvement on this limitation would be to include electromyographic assessments of the hamstring muscle involvement. Using these techniques, comparisons could also be made for muscle recruitment (not just muscle length) between participants with tight hamstrings and LBP and participants with tight hamstrings and no LBP to further strengthen the conclusions regarding the relationship between tight hamstrings and low back pain.
LIST OF REFERENCES


27. Hebert, C., Determination of the functional relationship between lumbar lordosis and pelvic tilt, in Orthopaedic Surgery and Biomedical Engineering. 2014, University of Tennessee Health Science Center: Memphis, TN.


APPENDIX A. INSTITUTIONAL REVIEW BOARD APPROVAL

October 24, 2013

Bill Mihalko, MD, PhD
UTHSC - COM - Orthopaedic Surgery
E226 Coleman College of Medicine Building
956 Court Avenue
Memphis, TN 38163-0000

Re: 13-02677-FB
Study Title: Determination of functional relationship between lumbar spine and pelvic plane alignment

Dear Dr. Mihalko:

The IRB has received your written acceptance of and/or responses dated 10/14/2013 and 10/23/2013 to the provisos outlined in our correspondence of 09/04/2013 and 10/21/2013 concerning the application for the above referenced project. The IRB has reviewed these materials and determined that they comply with proper consideration for the rights and welfare of human subjects and the regulatory requirements for the protection of human subjects. Therefore, this letter constitutes full approval by the IRB of your application Version 1.2 and the accompanying:

- Recruitment Flyer, dated 10/4/2013,
- Telephone and Email Script, dated 10/4/2013,

Approval of this study will be valid from 10/23/2013 to 09/04/2014.

The IRB has determined that the informed consent form, incorporating the authorization of subjects to use their protected health information in research, complies with the federal privacy regulations as specified in 45 CFR 160 and 45 CFR 164. In addition, in accord with 45 CFR 46.116(d), informed consent may be altered for your telephone and email screening, with the cover statement used in lieu of an informed consent interview. The requirement to secure a signed consent form is waived under 45 CFR 46.117(c)(2). Willingness of the subject to participate will constitute adequate documentation of consent.
In addition, the request for waiver of HIPAA authorization for the recruitment of subjects is approved.

In the event that subjects are to be recruited using solicitation materials, such as brochures, posters, web-based advertisements, etc., these materials must receive prior approval of the IRB. Any revisions in the approved application must also be submitted to and approved by the IRB prior to implementation. In addition, you are responsible for reporting any unanticipated serious adverse events or other problems involving risks to subjects or others in the manner required by the local IRB policy.

Finally, re-approval of your project is required by the IRB in accord with the conditions specified above. You may not continue the research study beyond the time or other limits specified unless you obtain prior written approval of the IRB.

Sincerely,

Margaret M. Sularin

Signature applied by Margaret M Sularin on 10/24/2013 09:08:30 AM CDT

Terrence F. Ackerman

Signature applied by Terrence F Ackerman on 10/24/2013 09:09:43 AM CDT

Margaret M. Sularin, LMSW, RD, LDN, CCRP
Regulatory Specialist
UTHSC IRB

Terrence F. Ackerman, Ph.D.
Chairman
UTHSC IRB
APPENDIX B.  ADDITIONAL GRAPHS

Below are graphs showing ankle, knee, and hip angles, moments, and powers throughout the full stair cycle. Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-1: Average Ankle Angles throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-2: Average Ankle Moments throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-3: Average Ankle Powers throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-4: Average Knee Angles throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-5: Average Knee Moments throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-6: Average Knee Powers throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-7: Average Hip Angles throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-8: Average Hip Moments throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
Figure B-9: Average Hip Powers throughout the Full Stair Cycle
Graphs in the left column represent average results from trials without the orthosis. Graphs in the right column represent average results from trials with the orthosis in place. The first row shows results during stance phase in stair ascent. The second row shows results during swing phase in stair ascent. Rows 3 and 4 show results during descent in stance and swing phase, respectively. Parameter comparisons in these graphs are made between those with tight hamstrings (seen in red) and normal hamstrings (seen in blue).
VITA

Dema Assaf was born in Cleveland, Ohio in 1989. She attended Bradley University in Peoria, Illinois, where she earned a Bachelor’s Degree in Mechanical Engineering in 2012. She then worked under Dr. William Mihalko as a graduate research assistant in the Joint Program in Biomedical Engineering at the University of Tennessee Health Science Center and the University of Memphis. She earned her Master’s Degree in Biomedical Engineering in August 2015.