QUANTIFYING SEGMENTAL SPINAL MOTION DURING ACTIVITIES OF DAILY LIVING

by

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DISSERTATION ABSTRACT

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Back pain is a very common musculoskeletal impairment in most Americans. Average annual occurrence of back pain is reported around 30% of the population and is the most common cause of activity limitation in people younger than 45 years old. Eighty percent of the back pain presents in the lumbar spine. Although this ailment is very prevalent in the American population, there is a lack of empirical evidence supporting the common clinical diagnosis and intervention back pain strategies. The frequency of back pain and the lack of treatment methods were the motivation for this investigation. It is important to better understand spine dynamics during ambulatory tasks of daily activities to identify possible biomechanical mechanisms underlying back pain.

Current biomechanical quantification methods for spine dynamics are either too invasive or not detailed enough to fully comprehend detailed spinal movement. Therefore, a non-invasive but detailed procedure to calculate spine dynamics was developed and tested. In this study, multi-segmented spine dynamics (kinematics and kinetics) were calculated during four activities of daily living (level walking (W), obstacle crossing (OC), stair ascent (SA) and stair descent (SD)).

Our findings suggested an in-vivo multi-segmented spine surface marker set is able to detect different and repeatable motion patterns during walking among various spinal

segments. The sacrum to lower lumbar (SLL) joint had the largest range of motion (ROM) when compared to the other more superior joints (lower lumbar to upper lumbar and upper lumbar to lower thoracic). Furthermore, SA task demonstrated more flexion ROM than both W and SD tasks. In addition to task influence, joints at different spine levels also demonstrated different ROMs, where SLL had a greater ROM than upper lumbar to lower thoracic (ULLT) in the transverse plane. Age was found to not significantly affect the segmental spinal ROM or peak angles. The vertical segmental joint reaction forces were different between tasks, where SD yielded larger vertical reaction forces than W.

Overall, findings from this dissertation work were able to show that a multi-segment spine marker system could be an effective tool in determining different spinal dynamics during various activities of daily living.

This dissertation includes unpublished co-authored material.

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CHAPTER I

INTRODUCTION

Back pain is a common musculoskeletal impairment among Americans (NINDS, 2011). The annual average occurrence of back pain is around 30% of the population, which ranges between 15% and 45% (Andersson, 1997). Moreover, approximately 65% of Americans will experience acute back pain annually, which is defined as the presence of pain for less than one year (Walker, 2000). More alarmingly, most of these individuals suffering from acute back pain do not completely recover (Croft, Macfarlane, Papageorgiou, Thomas, & Silman, 1998). Back pain is the most common cause of activity limitation in people younger than 45 years old (Praemer, Furnes, & Rice, 1992). It is the second most frequent reason for visits to the physician, the fifth-ranked cause of admission to hospital, and the third most common cause of surgical procedures (Hart, Deyo, & Cherkin, 1995; V. M. Taylor, Deyo, Cherkin, & Kreuter, 1994). Praemer et al. (1992) showed that of all the reported musculoskeletal impairments, back and spine injuries represented 57% of the total impairments in people 65 years or younger.

The most common back disorders afflict the lumbar spine and can affect up to 80% of individuals suffering from back pain (Kelsey & White, 1980). It has been estimated that close to 50 billion dollars are spent annually on the diagnosis and treatment of back pain disorders (NINDS, 2011). Due to the variety of back pain and the lack of consistency in prevention and treatment plans, the Healthcare Cost and Utilization Project (HCUP) listed seventeen different inpatient procedures dealing with the back/spine/spinal cord in some forms. They reported that, on average, back pain patients stayed in the hospital for approximately six days with an average medical cost of approximately $100,000 per patient (HCUP, 2010). This leads to an average in-patient hospital cost of 42 billion dollars annually for treating back pain patients in the US (HCUP, 2010). Furthermore, such expenditure increased by eight billion dollars between 2010 and 2011.

Back pain affects men and women similarly (NINDS, 2011). The most common age range to report back pain is between 30 and 50 years (NINDS, 2011). Some back pain symptoms, while causing inconvenience in performing activities of daily living, may not require a health specialist and can be treated with rest, stretching, exercise or diet changes (Bigos, Policy, Research, & Consultants, 1994). Back pain can be caused by numerous, more serious, ailments which require a visit to a health care professional. These may include; bulging (herniated) disc, sciatica, spinal degeneration, stenosis, fibromyalgia and spondylitis (NINDS, 2011).

Regardless of the occurrence and the possibility for back pain to progress from acute to chronic, there is a lack of empirical evidence supporting the commonly used clinical diagnosis and intervention strategies for back pain (Fritz, Delitto, & Erhard, 2003). The Agency for Health Care Policy and Research (AHCPR) in the United States followed this same doctrine and stated that people with back pain can return to low-stress aerobic activities after 2 weeks of symptoms (Bigos et al., 1994). The AHCPR later suggested that exercises to condition trunk muscles should be delayed at least 2 weeks post injury, while simultaneously suggesting that individuals who are recovering from back pain should return to work or their normal daily activities as soon as possible. The AHCPR also suggested if the back pain symptoms persist, further evaluation is needed. Clearly there is an inconsistency in how back pain should be treated.

The lack in unity in back pain treatment mainly stems from the inability to correctly identify effective intervention for each individual patient (Bouter, van Tulder, & Koes, 1998; Leboeuf-Yde, Lauritsen, & Lauritzen, 1997). In addition, classification processes for back pain vary greatly. Krause and Ragland (1994) proposed an eight-phase classification system for back pain, which takes into account biomedical, developmental, and social characteristics of work-disability. Several different questionnaires are used by physicians to assess back pain. The Roland and Morris disability questionnaire was developed in 1983 and has twenty-four questions asking about pain in various activities (Roland & Morris, 1983). A different questionnaire has ten sections that refer to activities of daily living that might be disrupted by back pain (Fairbank, Couper, Davies, & O'Brien, 1980). In the Fairbank questionnaire, patients are asked to choose from one of several sentences which best describes their pain during an activity of daily living. Providing yet another back pain diagnosis tool, Quebec Back Pain Disability Scale has twenty questions that are related to activities of daily living and the pain is assessed on a 0-5 likert scale (Kopec et al., 1995). Even after a study of which treatment method is the best concluded that returning to normal activities as soon as possible is advisable, no standard exists (Malmivaara et al., 1995). With so many options for the diagnosis of back pain, the lack of consistency in treatments is obvious.

To address the inconsistencies in the diagnosis and treatment of back pain, it is important to gain a better understanding of back mechanics during daily activities and how the introduction of these activities will influence back dynamics which may be related back pain. The study of back mechanics during activities of daily living will provide insight to how the back and spine respond to perturbations which are encountered daily. Therefore, the long-term goal of this study is to ultimately enhance our objective quantification of spinal motion that will result in evidence-based information for the development of better treatment and diagnosis techniques. The following literature review provides a summary of relevant research in spine mechanics. Discussed first are causations of back pain followed by narratives on common injury environments. Finally a discussion on previous spine biomechanics and the motion capture methods used will be presented.

Causation of Back Pain

Back pain can be a result of injury to back musculature, joints or discs (NINDS, 2011). It could be a result of small but repetitive or larger acute stresses (Walker, 2000). The small repetitive stressors could arise from the contact with the ground during walking or trunk muscle contraction (Chaffin, Andersson, & Martin, 2006; D. A. Winter, 2009). These forces are transferred from the foot through the hip to the back which are then combined with the forces generated from muscle contractions (Chaffin et al., 2006). An acute stress, which is generally the more studied mechanism of back pain, can result from blunt force trauma from the external environment. These acute types of forces can occur during an overexertion lifting task in the work place or in a car accident (Chaffin et al., 2006). Although this type of mechanism exists in the causation of back pain, it is not the focus of this dissertation investigation, which was undertaken to better understand the repetitive motions and forces that the back encounters on a daily basis. The long term aspirations are to provide empirical evidence that can be used in the development of consistent back pain diagnoses and treatments. To develop a method of better understanding of back mechanics, first, we must understand what tasks can lead to the musculature, joint or disc injuries responsible for back pain.

Back Injuries in Workplace and during Activities of Daily Living

Work related back injuries are important to understand when considering injuries associated with activities of daily living. With all the safety strategies imposed in today's work place, the documentation of injuries and the task which caused the injury is comprehensive. As many of the same repetitive tasks undertaken in the work place are similar to those accomplished during daily living (walking, negotiating stairs and obstacle crossing) some correlations can be drawn between the daily tasks and the job tasks resulting in back pain. One of the main causes of back pain is improper body mechanics, which is extremely common in the work place (NINDS, 2011). Some common work place injury mechanisms include heavy lifting, repeated motions and awkward stances. These improper mechanics could be more prevalent in daily home activities, as the work place training and educational programs are not available to people in everyday life. According to the 2010 Bureau of Labor Statistics (BLS) Injuries, Illness and Fatalities Report (IIF) overexertion in lifting and repetitive motions were one of the highest causes of injuries cited (BLS, 2010). Activities of daily living commonly have repetitive motions (e.g. walking and other ambulatory movements) which could possibly lead to back pain. This possibility is unknown due to the lack of documentation. For example, some studies were able to quantify low back motions and forces during walking, demonstrating that low back compressive stresses could range between 150 and 250% of body weight and occur at specific gait events, such as heel strike and toe off (Callaghan, Patla, & McGill, 1999; Cromwell, Schultz, Beck, & D., 1989; Whittle & Levine, 1999). Therefore, an investigation of the repetitive motions of the spine during activities of daily living could very well provide a more in-depth understanding of spinal dynamics that has not been previously reported.

The spine is a complex combination of bones, joints, ligaments, intervertebral discs and muscles responsible for support and movement of the body between the pelvis and upper extremity. An accurate understanding of how these components work together during activities of daily living is necessary to comprehend possible degenerative mechanisms that can lead to back pain thereby assisting in the development of suitable treatment options. Although these mechanisms are not well understood, the effects of back pain during daily activities are well documented. If an individual suffers from back pain, there could be a chain of post-injury reactions that can lead to disuse, disability and depression (Vlaeyen, Kole-Snijders, Boeren, & Van Eek, 1995). This is commonly known as a fear-avoidance cycle, which has been shown to be an active factor in the transition of acute back pain to chronic back pain and has been quantified using a questionnaire (Buer & Linton, 2002; Waddell, Newton, Henderson, Somerville, & Main, 1993). Additionally, patients with back pain tend to have a lower physical activity level, less standing time, lower step frequency and more lying time during the evening which leads to the resulting reduction in physical activity (Spenkelink, Hutten, Hermens, & Greitemann, 2002). Although there is no defined treatment for back pain, the most widely accepted intervention is to have the patient resume normal movement and daily activities as soon as possible (Bigos et al., 1994). With the obvious issues that back pain inflicts on people's daily lives, attempts must be made to better understand the mechanisms which cause this ailment. This is important for the development of better treatment procedures, as well as preventative methods in place to reduce the number of people that suffer from back pain. This work strives to take the first step in that process by developing an in-vivo motion analysis protocol which allows an examination of detailed spinal motion during ambulatory tasks of activities of daily living, which would provide insightful biomechanical evidence to demonstrate how a back injury occurs or affect the performance of daily activities.

Spine Biomechanics during Lifting and Locomotion

Due to the high incidence of injury in the work place, most of the back pain: research is focused on one of the acute causes of back pain, overexertion in lifting. This research focused on the forces and moments applied to the junction between the 5th lumbar and 1st sacral vertebrae (L5/S1) or 4th lumbar and 5th lumbar 5 (L4/L5) joints (van Dieen, Hoozemans, & Toussaint, 1999). One of the earlier models was developed to estimate these forces and included the erector spinae muscles and took into consideration the rate at which the load was accelerated during lifting (Park and Chaffin (1974). It was one of the first studies to show that forces on the erector spinae muscles and the lumbosacral disc can be as much as 50% higher when using the recommended “straight back, bent knees” if the load which is lifted is initially too far away from the spine. Although this model is still used today, it has several limitations, such as: modeling the entire back as one rigid segment and focusing in the sagittal plane. With better technology, lifting lumbar spine dynamic back models became more popular and more elaborate. For example, Freivalds, Chaffin, Garg, and Lee (1984) were able to detail the implementation of a dynamic biomechanical model using actual segment motion data. Additionally, an extensive anatomical detail has been incorporated to a three-dimensional musculo-ligamentous-skeletal system (McGill & Norman, 1986). In addition, to address the amount of back injuries which occur in the workplace from lifting, the National Institute of Occupational Safety and Health (NIOSH) developed a lifting equation (1994) which accounts for several variables such as coupling to the lifted object, how high the object needs to be lifted, how much the individual must twist during the lift and reports a recommended maximum weight of lift. The NIOSH equation is still widely used by ergonomists to assess lifting tasks (Waters, Putz-Anderson, Garg, Safety, & Health, 1994). The NIOSH equation has a frequency component (Waters et al., 1994) which could be applied to non-ambulatory activities of daily living. Although no direct use of the NIOSH model can be found for this study, due to the lack of ambulatory component, the application of this equation to other activity of daily living should be considered. Although this lifting research is important, more attention should be given to what occurs on a daily basis during dynamic movement. The repetitive motions of gait and how the spine reacts to these motions are not well understood. Many of the lifting studies have reported kinetic responses, such as joint reaction forces and moments, to the spine. Several studies have discussed kinematic responses during lifting. These discussions are generally focused on how to position the back in order to reduce possible injuries during lifting. Kinematic considerations tend to focus on the lordosis of the lumbar spine during lifting and how lifting influences alters lumbar lordosis as well as how changes in the lordosis during lifting can alter erector spinae activation (Hwang & Kim, 2009; Kahrizi, Parnianpour, Firoozabadi, Kasemnejad, & Karimi, 2007; McGill, Hughson, & Parks, 2000). Although lumbar lordosis could be measured with a detailed spine model, it is the not focus of this work. This research is focused on how ambulatory tasks will affect spinal motion. It is important to mention lifting studies when proposing a new spine quantification method, as a great majority of research has been done with this task.

Gait studies have traditionally focused on the lower extremity, investigating kinematics and kinetics in the ankle, knee and hip joints (Collins, Ghoussayni, Ewins, & Kent, 2009; Kadaba, Ramakrishnan, & Wootten, 1990; Ounpuu, Gage, & Davis, 1991; D. A. Winter, 2009). Several studies have targeted spine motion during gait which would focus solely on the lumbar spine and/or define entire regions of the spine using one rigid segment (Callaghan et al., 1999; Crosbie, Vachalathiti, & Smith, 1997b; Goh, Thambyah, & Bose, 1998; N. Taylor, Evans, & Goldie, 1996; Vogt & Banzer, 1999; Vogt, Pfeifer, Portscher, & Banzer, 2001; Whittle & Levine, 1999; Yamamoto, Panjabi, Crisco, & Oxland, 1989). The use of large segments to describe spinal motion does not provide detailed segmental movement, and therefore biomechanical information regarding the spinal motion is compromised. A more detailed movement description is needed, considering the spine has three degrees of freedom and has three regions above the hips (Gray, 2009). Each level of the spine is comprised of vertebrae, with different facet joints allowing specialized movement for each (White & Panjabi, 1990). The spinal facet joints are oriented in such a manner that each region of the spine has a particular movement they are premeditated to do. For example, the lumbar segment is specialized to perform flexion and extension, the thoracic facets are specialized for lateral bending but allows for axial rotation, and the facet joints in the cervical region are tri-axial (Schmitt, Niederer, & Walz, 2004). It is therefore important to incorporate as many degrees of freedom into a spine model. This will allow for a greater understanding of how each segment moves during various dynamic conditions. The knowledge gained from a complete quantitative method of the spine will be most useful in developing preventative therapies and better diagnostic techniques which can be used in the clinic.

In-Vivo Motion Capture of Spinal Motion

The ability to quantify movement of each spinal segment is a crucial step to better investigate biomechanical interactions between different spinal segments during dynamic tasks. Several different methods have been used to capture back motion, including a magnetic tracking device (P. J. Rowe, 1996; P. Rowe & White, 1996), electric goniometry (Marras et al., 1993) and optical tracking with surface markers (Crosbie et al., 1997b). A few studies have utilized finite element analysis to quantify the movement and forces on the vertebral column (M. Lee, Kelly, & Steven 1995). While other researchers have used bone pins to provide accurate measurements of movement of the lumbar spine during uni-planar movements (Rozumalski et al., 2008). Additionally, electromyography (EMG) and ultrasound have also been used to examine the forces in the spine(Callaghan et al., 1999; Vogt & Banzer, 1999).

Previous studies have investigated back motion during gait with the utilization of a single or large spinal segment, defined superiorly by the right and left shoulder markers and inferiorly by a sacral marker (Rowe 1996; Crosbie, Vachalathiti et al. 1997; Whittle and Levine 1999; Vogt, Pfeifer et al. 2001; Riley, Paolini et al. 2007(Crosbie et al., 1997b; Riley, Paolini, Della Croce, Paylo, & Kerrigan, 2007; P. J. Rowe, 1996; Vogt et al., 2001; Whittle & Levine, 1999). Riley et al. (2007) compared spinal kinematics between over ground and treadmill walking. The researchers utilized the Plug-in-Gait model, defined superiorly by the right and left shoulder markers and inferiorly by a sacral marker (Vicon Peak, Lake Forest, CA, USA). Although the study found differences in spine flexion/extension between over ground and treadmill walking, this model defined the trunk as one large segment. Similarly, Crosbie et al. (1997b) reported spinal patterns between lumbar, upper and lower trunk and pelvis segments during gait. Each segment consisted of only three markers to define each of the lumbar and thoracic regions. This model showed consistent and repeatable patterns; however by using such large segments (e.g. 12 vertebrae in thoracic and 5 vertebrae in the lumbar), for each region of the spine the finer details of the segmental motion may have been missed.

The complexity of the spine is obvious and it is clear a detailed quantification of spinal motion is necessary, as there is a lack of detailed spine models. The one segment (or large regional segment) definition of the spine neglects the seventeen bones and over thirty joints in the lumbar and thoracic spines. Modeling the spine in this single segment manner does not seem practical, as movement between individual vertebra has been previously reported (White & Panjabi, 1990). Consequently, modeling the spine using a one segment approach neglects these motions that exist and contribute to the overall motion of the spine. The overall goal of this study is to develop an in-vivo marker set to quantify detailed motion of the spine. The results from the current study will not have a direct effect on clinical procedure, however, in the long term, the proposed procedure can possibility help in the development of screening techniques to determine if individuals are predisposed to developing back pain. This would be accomplished by finding correlations between easily observable clinical measures (step-width, step-length, cadence, etc.) and spine dynamics. Rozumalski et al. (2008) inserted bone pins into individual spinous processes and track individual motion of each vertebrae. In conjunction with the work of White and Panjabi (1990), this study also found motion between individual vertebral bodies, thus solidifying the need for acquiring finer motion of the spine by using smaller segments. Although this method provided accurate tracking of each vertebra, it employed invasive surgical procedures for bone pin placements. This current study attempted to develop an in-vivo and non-invasive methodological precedence for more accurate quantification of the spine movement. To do this, we defined smaller spinal segments using palpable spinous processes and tracked their motion during dynamic tasks using surface markers. With the accuracy and precision of the optical tracking systems currently available, it is feasible to detect motion exhibited by a smaller spinal segment.

Most investigations on spine motion have focused on lifting (Dempsey, 2002; Faber, Kingma, Bakker, & van Dieen, 2009; Freivalds et al., 1984; Hwang & Kim, 2009; McGill & Norman, 1986; Mitnitski, Yahia, Newman, Gracovetsky, & Feldman, 1998; Park & Chaffin, 1974; van Dieen et al., 1999). Studies have examined spine movements during level walking (Feipel, De Mesmaeker, Klein, & Rooze, 2001; Goh et al., 1998; N. Taylor et al., 1996; Vogt & Banzer, 1999). Many studies have investigated non-level walking, which could introduce biomechanical deviations or perturbations to the lower extremities. Obstacle crossing has been examined during level walking and was found to perturb the center of mass (COM) motion in the frontal plane (Chou, Kaufman, Brey, & Draganich, 2001). Age-related modification in the COM motion was also observed in the anterior-posterior direction (Hahn & Chou, 2004). Moreover, walking speed can affect lower body dynamics, such as moments, angular velocities and temporal-distance parameters (Draganich & Kuo, 2004). Although the COM is located within the trunk, it is not a direct measure of spine motion during obstacle crossing. Some research has incorporated direct spine measures during obstacle crossing by investigating kyphosis in individuals with osteoporosis (Sinaki, Brey, Hughes, Larson, & Kaufman, 2005). This study reported that back extensor and all lower extremity muscle groups were weaker in individuals who had osteoporotic related kyphosis than individuals who did not. Additionally, Hahn and Chou (2003) discussed how obstacle crossing could influence individual segment motion and determine instability in the elderly. It was reported that the trunk segment angles would change as a result of the increasing height of the obstacle. Although this study was able to show that obstacle crossing could affect the upper extremity, the spine was defined with the one large trunk segment previously mentioned, thus negating all the individual motions of the adjacent vertebrae.

Scannell and McGill (2003) studied angle differences between L1 and S1 using inclinometer during sitting, standing, and walking and reported lordosis can be changed following training. Additionally, range of motion (ROM) of the trunk has been reported during different activities of daily living, where differences were found between gait and stair ascent and descent in all planes of motion (Krebs, Wong, Jevsevar, Riley, & Hodge, 1992). More recently, a magnetic tracking device was placed at the 12th thoracic (T12) and 1st sacral (S1) spinous process, to measure spinal motion during stair walking (J. K. Lee & Park, 2011). Clear differences were reported in spinal motion between the level walking and staircase walking conditions, particularly in regards to the motion pattern and ROM of the flexion/extension and lateral bending of the spine. Similar to previous studies, the back and spine were defined as one large segment. Although the literature is limited, it is promising that differences have been reported previously in spine motion in response to different tasks.

The previous studies reported differences in spinal kinematics during obstacle crossing and stair navigation. Many of these results relied on large segment definitions of the spine and trunk for analysis. The large segment approach for quantifying spine motion ignores the motion that occurs between the seventeen adjacent vertebrae and the over twenty-five joints in the lumbar and thoracic spine. Furthermore motion has been recorded between the individual vertebrae, thereby it does not make sense to neglect motion vertebral motion by modeling the back and spine as a single segment (White & Panjabi, 1990). The ability to more completely capture the motions produced by the spine during ambulation will provide a much more complete biomechanical understanding of spine kinematics. A more complete biomechanical understanding of the spine could provide new information regarding possible mechanisms which lead to back pain, therefore not only leading to better treatment procedures, but also new preventative techniques. One of the purposes of this work is to develop a marker set with smaller segments of the spine to help better understand the motion of the vertebrae, not just the junction of the pelvis and the trunk as the single segment back model describes.

Aging and Spinal Motion

Gait patterns, such as stride width and length, in the elderly have been shown to adjust to help with the loss of balance control (Fiatarone & Evans, 1993). There have been reported decreases in stride length and increases in stride frequency in older populations (JudgeRoy, Davis, & Õunpuu, 1996; Menz, Lord, & Fitzpatrick, 2003). Additionally, an increase in step width has been found in individuals of advanced age (Maki, 1997). In addition to changes in the spatiotemporal parameters, age related changes are present in the anterior/posterior range of motion of the center of mass during obstacle crossing (Hahn & Chou, 2004), medio-lateral stability (Schrager, Kelly, Price, Ferrucci, & Shumway-Cook, 2008) and during dual task situations (Hollman, Kovash, Kubik, & Linbo, 2007). This indicates the aging process has a significant effect on an individual's motion characteristics.

Aging of an individual has been shown to affect the trunk and spine motion. Gill et al. (2001) was able to detect differences between old and young individuals by the range of angular sway and velocity in the trunk. Different trunk acceleration characteristics between young and older individuals were also reported (J. J. Kavanagh, Barrett, & Morrison, 2005; J. Kavanagh, Barrett, & Morrison, 2004). Aging studies have historically focused on the on the lower extremity (Hahn & Chou, 2004) or used questionnaires (Gloth III, Walston, Meyer, & Pearson, 1995; Kawashima, Motohashi, & Fujishima, 2004) to assess aging. How aging affects the dynamics the spine have not been well reported, nor has any detail been given to the spine in those studies. A better biomechanical understanding of how the effect of aging will induce changes into the spine motion is needed. This will be done by introducing and developing detailed spine model. Information gained from studies such as these will provide new insights regarding the movement of the spine as a result of aging. Ideally, this could lead to new clinical procedures in both the diagnosis and treatment of back issues in the elderly.

Joint Reaction Forces

Lower extremity joint reaction forces have been commonly examined during gait analysis (D. Winter, 2005). Differences have been observed in lower extremity reaction forces during different activities of daily living (Costigan, Deluzio, & Wyss, 2002; Draganich & Kuo, 2004; Hahn & Chou, 2004; Protopapadaki, Drechsler, Cramp, Coutts, & Scott, 2007). It is expected that due to the reported joint reaction force changes in the lower extremity from various activities of daily living, there will be observable changes in the segmented spine joint reaction forces.

Joint reaction forces have also been reported in the spine. Khoo, Goh, and Bose (1995) developed a method to calculate the forces in the lumbosacral joint during normal gait. Other models have used a similar large segment to define the back and its inertial properties (Park & Chaffin, 1974). Estimations of lumbosacral forces have been investigated during backpack loads (Goh et al., 1998; Hong & Cheung, 2003), during over ground walking (Callaghan et al., 1999) as well as on a treadmill (Feipel et al., 2001). Some direct measurements have also been recorded with implanted wireless spinal fixations while data were recorded during walking (Rohlmann, Bergmann, & Graichen, 1997). The changes reported in the lower extremity kinetics during various tasks, in conjunction with the stated differences at the lumbosacral joint, suggests that the joint reaction forces will be altered by numerous tasks at distinct spinal levels. The understanding of the joint reaction forces for various levels of the spine is imperative in developing a full biomechanical perceptive of spinal dynamics during ambulatory activities of daily living. This information will aid in the understanding of possible degenerative mechanisms in the spine thereby helping to develop appropriate treatment and preventative options.

Empirical Evidence to Clinical Application

Back pain has been presented as a serious issue that numerous people deal with on a daily basis. The frequency and financial burden back pain is placing on the health care system is alarming and the lack and inconsistency of treatments is astonishing. Although this research will not be able to directly address back pain, the motivation for this study was to develop a procedure that will be able to provide and report accurate knowledge of the dynamics in the thoracic and lumbar spine during ambulatory activities of daily living. A precise understanding of spinal dynamics will help to uncover and understand mechanisms which may lead to pain in the back, thereby allowing for future development of treatment and diagnosis options.

To do this, a new in-vivo method for quantifying spine dynamics is proposed. This procedure will address the limitations in the previous spine definitions (e.g. large or single segment spine definitions) as well as investigate how various ambulatory activities of daily living, in conjunction with the effects of aging, will alter segmented spine motion and joint reaction forces. To address the large segment issue that has predominantly been used to describe spine motion, smaller segments (consisting of only 3 vertebrae each) will be rigidly defined and kinematics and kinetics of these adjacent segments will be reported. Again, the impetus of this work is to provide empirical evidence that can eventually be used to improve the treatment of back pain.

Overall Goals and Specific Aims

The previous summaries have shown back pain is a major problem. There are inconsistencies in the treatment and diagnosis of this aliment. To better understand back pain, the dynamics of the healthy back and spine must be understood. This research strives to enhance the understanding of back biomechanics by providing knowledge of the underlying dynamic mechanisms that are present in the back which may lead to back pain. The results of this study is intended to provide a basis for which further investigation can lead to the development of better treatments and diagnosis procedures for back pain. This research has the following long-term goals:

1. To advance our understanding of spine mechanics during activities of daily living with the implementation of an in-vivo marker-based protocol
2. To enhance our investigation of underlying biomechanical mechanisms leading to back pain, and
3. To provide evidence-based knowledge that allows for the development of effective clinical regiments of interventions to alleviate and prevent back pain.

Within the context of these overall long term objectives, three specific aims were proposed:

(1) Develop and validate an in-vivo marker-based motion analysis protocol which could provide reliable kinematic quantification of several different spinal segments during locomotion.

Hypothesis: specific spinal segments will present unique motion patterns which have not previously been recorded and will differ from the commonly used single segment spine definition, thus providing more insight of the control of the spine.

(2) Examine the effect of individual activities of daily living, e.g., level walking, obstacle crossing, stair ascent and descent, on kinematics of individuals spinal segments in young adults.

Hypothesis: Different activities of daily living will exhibit unique kinematics at corresponding spinal segments.

(3) *Investigate whether individuals in consecutive age* groups *(20-29, 30-39, 40-49, 50-59 years) will display altered spinal kinematics during level walking and stair descent.*

Hypothesis: Individuals in different age groups will exhibit kinematics changes at the specific spinal segments during level walking and stair descent.

(4) Explore the feasibility to quantify the joint reaction forces at various spinal joints during different activities of daily living in young adults.

Hypothesis: unique activities of daily living will produce physiological reasonable joint reaction forces at specific joints of the spine.

Summary

The purpose of this research was to develop and test the feasibility of an in-vivo motion analysis protocol that allows for the examination and quantification of segmental spine motion during walking, obstacle crossing, and stair ascent/descent in different age populations. The information from this research is intended to expand the traditional knowledge of gait analysis with the inclusion of detailed spinal movement and allow for a better understanding of the dynamic coupling between the upper body and lower extremities. Such methodology enhancement could yield important biomechanical data such as the dynamic responses of the spine during activities of daily living. This information could be paramount for furthering the investigation of back pain and eventually lead to the development of more effective treatment and screening techniques. New treatment techniques are needed as the procedures used by physical therapists to treat back pain are generally not based on empirical evidence.

To evaluate such an in-vivo motion analysis protocol for the examination of detailed spinal motion during activities of daily living, several biomechanical issues need be addressed. First, segmentation of the spine must be physiologically and anatomically meaningful, as well as methodologically feasible. The proposed segmentation should result in repeatable kinematic measurements and provide a finer resolution to differentiate movement from opposing spinal sections. Secondly, similar to the lower extremity joint kinematics, baseline measurements of individual spinal segments during normal walking need to be established. Furthermore, it is also important to quantify the extent of changes in spinal motion due to aging as well as specific activities of daily living.

This research project was designed with two primary components. First, there was an experimental component to collect anthropometric data, body motion data and ground reaction force data during various activities of daily living (level walking, obstacle crossing, stair ascent & stair descent) from four healthy age populations with a maximum of 12 subjects in each group: 20-29, 30-39, 40-49 & 50-59. Secondly, there was an in-vivo marker set development which was utilized to examine the dynamics (kinematics & kinetics) expressed by the spine during multiple activities of daily living. The information gained from this research increases the understanding of the biomechanics of the spine during activities of daily living, which could help identify the underlying mechanisms associated with back pain.

Flow of Dissertation

This dissertation is structured in a journal format. Following the general review of literature (Chapter I), Chapters II through V represent individual manuscripts (co-authored materials by Dr.Li-Shan Chou) that have been published, or are in various stages of submission/revision to peer-reviewed scientific journals.

Following the general introduction and the review of literature (Chapter I); Chapter II describes the development and repeatability of the marker set used to quantify detailed spinal motion. This marker set is tested under a level walking condition and is compared to the motion of large spinal segments. The following chapter (III) details the differences which exist between distinct activities of daily living (ADL's), such as level walking, obstacle crossing, stair ascent and stair descent. Specifically, this chapter will show how unique ADL's can alter movement patterns within the spine in a young population.

The fourth chapter (IV) explores the age-related effect on spinal kinematics. In this chapter the motion change is described by the changes in range of motion (ROM) and peak excursion. Chapter V describes the observed spine joint reaction forces during four ADL's. This chapter will show that forces exerted on the spine can be significantly altered depending on a person's activity.

Finally, a general summary is provided in Chapter VI. This chapter will review the individual experiments, and include conclusions drawn from the major findings of each. Strengths and weaknesses of the studies are then discussed including suggestions for future studies. Appendices are provided prior to the bibliography, showing the informed consent form and a MATLAB® script for the various biomechanical outcomes which was used in the development of the mathematical calculation of the spinal dynamics.

CHAPTER II

QUANTIFICATION OF MULTI-SEGMENTAL SPINE MOVEMENT DURING GAIT

This chapter was developed by Li-Shan Chou, Ph.D. & Scott P. Breloff. Dr. Chou contributed substantially to this work participating in the development of methodologies and providing invaluable critiques and substantial editing advice. Scott P. Breloff was the primary contributor to the development of the protocol, data collection, data analysis and did the writing.

Introduction

Back pain is a serious problem that occurs annually in approximately 30% of the population (NINDS, 2011). Back pain can be the result of musculature, joint or disc injury (NINDS, 2011). Back pain is the second leading aliment which requires a visit to the physician, the third most common pathology for surgery and fifth leading cause in individual admissions to the hospitals (Hart et al., 1995; V. M. Taylor et al., 1994). Pain in the back region poses a major finaical burden to the healhcare system in America. In 2011, the estimated costs assoicated with back pain were approximately 42 billion dollars (HCUP, 2010). One year later the NINDS (2011) estimated healthcare costs assoicated with back pain to be close to 50 billion dollars. Back pain is clearly a burdening ailment in its frequency and financial costs.

Despite the burden back pain presents, there is no standardized diagonsis and treatment. Surprisingly, most of the interventions provided by clinications are not supported by emperical evidence (Bigos et al., 1994; Fritz et al., 2003). One possibility for discrepancies in back pain treatments could be due to the inability to correctly identify the most useful treatment type for each individual patient (Leboeuf-Yde et al., 1997). Treatment is usually prescribed based on responses to a questionnaire completed by the patient. These questionnaires are not standardized, and several different versions are commonly used (Kopec et al., 1995; Krause & Ragland, 1994; Roland & Morris, 1983). One study synthesized these possible diagnosis and treatment methods but was unable to determine which treatment method was the best (Malmivaara et al., 1995). Therefore the development of a procedure that could provide a detailed examination of back motion during dynamic movement would be beneficial for the identification of biomechanical factors contributing to back pain and eventually enhance the healthcare procedures of this ailment.

It has been documented that individuals who are suffering from acute back pain generally experience a fear avoidance cycle, which has been shown to lead to a chronic back pain (Buer & Linton, 2002; Vlaeyen et al., 1995). This transition of back pain classification occurs because patients were reported to have a lower physical activity level, less standing time, lower step frequency and more lying time during the evening (Spenkelink et al., 2002). To prevent this transition from acute to chronic back pain, new diagnostic and treatment protocols must be developed and deployed. The development of better techniques depends on an accurate examination of spine motion during dynamic activities. It is therefore necessary to develop a method which will provide a complete understanding of spine motion during gait. A complete understanding of spine motion will present new insights to possible degenerative mechanisms that could lead to back pain. A precise knowledge of these mechanisms can be used to develop new treatment protocols as well as preventative procedures to address back pain.

The area between the 7th cervical and 5th lumbar vertebrae is commonly classified as the 'trunk' segment. This segment has been commonly used to describe back/spine motion during a typical gait analysis and is defined superiorly by the left and right shoulder markers and inferiorly by a marker placed midway between the left and right posterior superior iliac spine (Vogt & Banzer, 1999). This single segment approach melds together the lumbar and thoracic segments of the spine and can only describe limited motion exhibited by the lumbar and thoracic spine, given that the spine consists of seventeen bones with over thirty joints (Gilroy, MacPherson, Ross, & Schuenke, 2008). Some improvements for recording spine motion were made when a segmental definition of the spine was used to measure spinal motion during gait. Although the spine was defined using a segmented approach, these segments defined each of the spine regions (e.g. lumbar and thoracic) with one segment (Crosbie et al., 1997b). These procedures which use a single segment to define and quantify back motion are not detailed enough as individual motion between vertebrae has been reported previously using cadaveric and radiograph studies (White & Panjabi, 1990). Additionally, these studies reported only simple single plane motion (flexion/extension, lateral bending and axial rotation) and did not report movement of the spine during complex motions such as walking. Other studies have also reported individual motion of the vertebral bodies, but are much too invasive to be used in a large scale clinical setting, as the procedure required invasive surgical procedures to implant bone pins (Rozumalski et al., 2008). Based on the current spinal quantification techniques and knowledge, it is suggested that an in-vivo marker set be developed that is both detailed enough to record the motion of the lumbar and thoracic regions of the spine during ambulation and have the ability to be applied to a large cohort of individuals in the clinical setting.

Quantifying the motion between adjacent spinal segments will allow for the better examination of the complexity of spine motion during gait. This knowledge could provide insightful information to gain comprehensive understanding of possible injury mechanisms in the spine. Therefore, the purpose of this study was to implement and validate an in-vivo marker-based motion analysis protocol that could provide reliable kinematic quantification of different spinal segments during locomotion. It is hypothesized that different spinal segments will present different motion patterns during selected activities of daily living, and these different patterns could be reliably quantified using a surface marker-based motion capture method.

Methods

Subjects

Ten healthy young adults (5 males/5 females; mean age: 26.8±3.8 years, mean height: 180.5±27.7 cm, and mean body mass: 67.7±11.6 kg) were recruited from the university community to participate in the study. No subjects had a history or clinical evidence of neurological, musculoskeletal or other medical conditions affecting gait performance, such as stroke, head trauma, neurological disease (i.e. Parkinson's, diabetic neuropathy), visual impairment uncorrectable by lenses and dementia. All subjects reviewed and signed an informed consent approved by the Institutional Review Board prior to their study participation.

Experimental Protocol

Subjects were tested on two separate days with identical procedures. They completed three different tasks; anterior bending (flexion) and lateral bending movement while seated and level ground walking. For the seated tests subjects sat on an adjustable bench. A golf ball with reflective tape hung from the ceiling was used as a target for the subject to perform the bending tasks. The ball was placed at a location which formed a 45˚ anterior or lateral bending angle from the neutral seated position of the subject (Preuss & Popovic, 2010). The angle of bending was determined using a gonimeter. The axis of the gonimeter was placed on the greater trochanter of the femur for the forward flexion bending. The superior anatomical landmark was the shoulder for one of the gonimeter arms and the anatomical landmark was along the femur. During the lateral bending trials the axis of the gonimeter was places at the mid-point between the left and right posterior superior iliac spines. One gonimeter arm was lined up with the 7th cervical vertebrae and the other was placed along an extended line drawn by using the left and right posterior superior iliac spine. Subjects performed five trials for each bending condition and were then asked to perform level walking. Data collected from 5 successful trials were used for analysis.

Experimental Instrument

Prior to testing, subjects were asked to wear spandex shorts (with no shirt for men & dance leotard with open back for women, Figure 1A). Twenty-two retro-reflective markers (14mm in diameter) were placed on the back of each subject. Whole body motion data were collected with a ten-camera motion analysis system (Motion Analysis

Corporation, Santa Rosa, CA). Eight of these markers were placed directly on the palpated spinous processes of the following vertebrae; Cervical 7 (C7), Thoracic 3 (T3), Thoracic 6 (T6), Thoracic 9 (T9), Thoracic 12 (T12), Lumbar 3 (L3), Sacrum 1(S1) & Sacrum 5 (S5). The S1 marker is placed slightly above the 1st sacral spinous process. This allowed for the creation of a joint between the sacrum and pelvis segment and the lower lumbar segment. Two makers were placed on the left and right posterior superior iliac spine (PSIS), and the remaining markers were placed 50mm to the left and right of the spinous process markers to simulate the location of the left and right transverse process of the vertebrae (except for S1 and S5). This marker set (Figure 1B-1D) is similar to one described previously (Preuss & Popovic, 2010). In addition, kinematic data of the lower extremities during walking were also collected (Hahn and Chou, 2004)

Three force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were placed in series and embedded level into the laboratory floor. The first two force plates were immediately adjacent to each other, while the third plate was separated by a distance of 15 cm. This setup was to accommodate subjects walking with different step lengths. Gait events such as heel strike (HS) and toe off (TO) were detected using the vertical ground reaction force (GRFv). Heel strike was identified as the instant when the GRFv was greater than 10% of the maximum GRFv, and toe off was determined as the GRFv was less than 10% of the maximum GRFv (Ghoussayni, Stevens, Durham, & Ewins, 2004; Hreljac & Marshall, 2000; Mickelborough, Van Der Linden, Richards, & Ennos, 2000).

Definition of the Inter-Segment Angles

Six joint angles were defined to describe the relative motion between two adjacent spinal segments as: Sacrum-to-Lower Lumbar (SLL), Lower Lumbar-to-Upper Lumbar (LLUL), Upper Lumbar-to-Lower Thorax (ULLT), Lower Thorax-to-Middle Lower Thorax (LTMLT), Middle Lower Thorax-to-Middle Upper Thorax (MLTMUT), and Middle Upper Thorax-to-Upper Thorax (MUTUT), Figure 1D. Two additional angles were calculated; similar to the one reported previously using a large spinal segment: Sacrum-to-Upper Thorax (SUT) (between the most distal segment [sacrum] and most proximal [upper thorax] segment) and Sacrum-to-C7 (SC7).

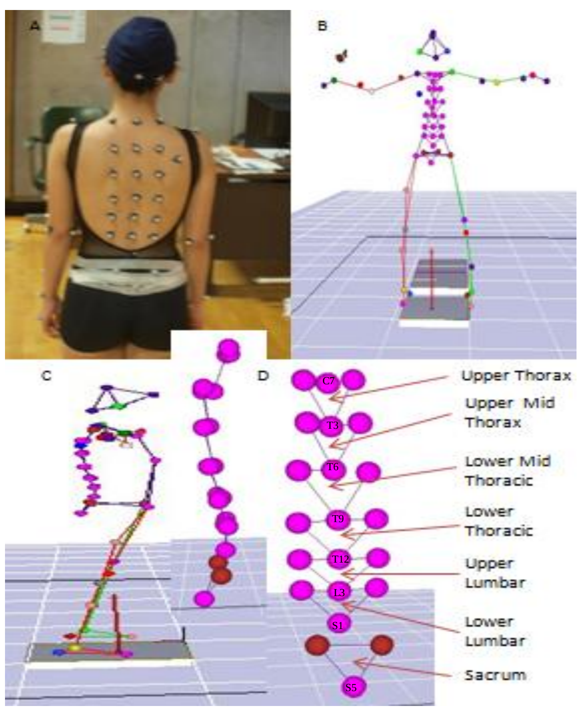


Figure 1. Example marker set (A) A female subject wearing the open back leotard with the marker set attached (B) Full body marker motion capture marker set shown in anterior view (C) Full body marker motion capture marker set shown in sagittal view. Blow up window shows only spine marker set (D) Close up of marker set in frontal plane with included segments names.

This segment was the relative motion between the sacrum and a segment defined by the most caudal back maker [S1] and the two shoulder markers (Jian, Winter, Ishac, & Gilchrist, 1993; Kadaba et al., 1990). Cardan angles were used to describe relative rotations between two adjacent segments. Axes of rotation were defined as followings: the X axis is pointing to the right (from posterior view); the Y axis is pointing anteriorly, and the Z axis is pointing upwards (Preuss & Popovic, 2010). The first rotation (e’) took place about the X-axis and denoted flexion and extension. The second rotation (e’’) was about the Y-axis and denoted lateral bending. The third rotation (e’’’) was about the Z-axis and denoted the axial rotation.

Data Analysis

Joint kinematic data of the bending conditions were analyzed from the beginning of upright sitting posture through a defined full range of motion (45^0) and return, to the beginning posture. For walking trials, the data were analyzed for a gait cycle (between ipsilateral heel strikes). A MATLAB® (Mathworks, Natick, MA) program was used to calculate all joint angles and the associated range of motion for each task. All angles were referenced to the values obtained during a static trial by subtracting the static pose angles from the angles obtained from dynamic trials. The normalization to the static pose was done so only motion of the spine during walking was discussed. Static pose for the seated trials required the subjects to sit comfortably upright on the piano bend with their hands on their lap. The walking trial static pose required the subjects to stand with their feet approximately should with apart, arms full extended and abducted to ninety degrees and look straight ahead. Joint angles were only calculated for the plane of motion in seated bending tasks (anterior or lateral bending), however for the walking condition, angles about all three anatomical axes were examined.

With the lack of gold standards, the repeatability and reliability were assessed to determine the feasibility of this in-vivo motion capture of spinal motion. Repeatability was assessed with the range of motion (ROM) and coefficient of multiple correlation (CMC). It has been reported that different spinal motion patterns could exist in healthy individuals, which increases the difficulty when comparing motions among different subjects (Gatton & Pearcy, 1999). The ROM has been suggested as a viable measure to assess repeatability of spine motion (Gatton & Pearcy, 1999). T-tests were performed to compare the mean of the ROM between the two testing days. In addition, an intraclass correlation coefficient (ICC) was calculated to determine the reproducibility of the spine angle ROM between the two testing days for all three task conditions. The statistical software package PASW (version 18, IBM., New York, NY) was used for all statistical analyses.

CMC has been used to determine the repeatability of joint angles during a gait cycle (Ferrari, Cutti, & Cappello, 2010; Kadaba et al., 1990). CMC values were calculated for each of the eight spinal joint angles to assess both within- and between-day repeatability for all three testing conditions. During initial data analysis, it was observed that an overall curve shift could exist between joint angles obtained from different days. This offset could be due to subtle differences in marker placements. As this current investigation focused on the overall pattern and range of motion of theses spinal joint angles, CMC values were calculated based on the normalized curves that described angular deviations from the value obtained at heel strike. A MATLAB® (Mathworks, Natick, MA) program was developed and used to calculate the CMC values based on the methods reported previously (Ferrari et al., 2010).

Results

Seated Bending Tasks

An initial visual comparison of the overall bending angles with those reported by Preuss and Popovic (2010) revealed an agreement with the current study (Figure 2). No significant between-day differences were detected in ROM for any of the spinal joints during either the anterior or lateral bending task ($p \geq 0.14$; Table 1), which demonstrated an acceptable repeatability of using the proposed marker set to quantify inter-segmental motion of the spine.

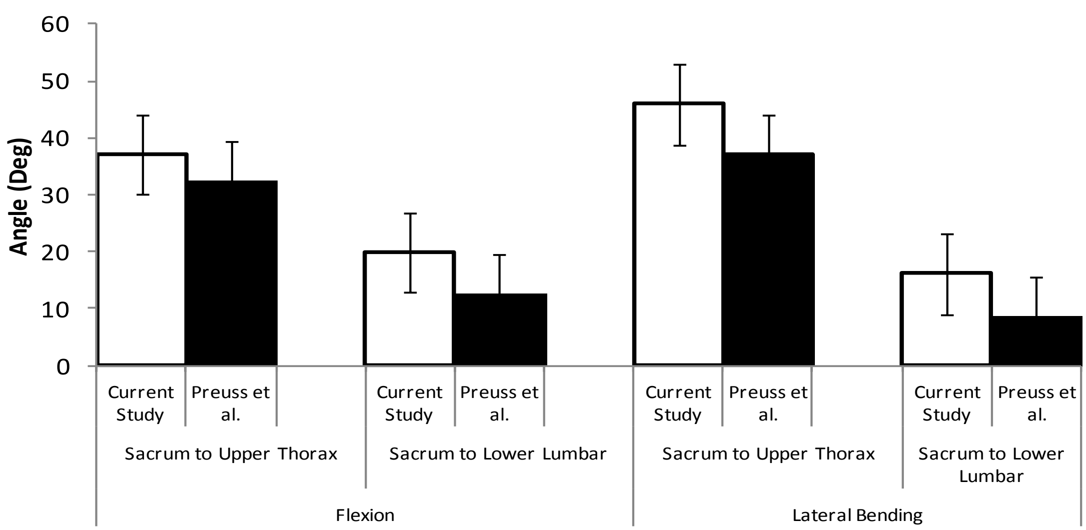


Figure 2. A graphical comparison between current study values and values reported by a previous study (Preuss & Popovic, 2010).

Table 1: Ranges of motion for anterior and lateral bending conditions

Angle^0	Lateral Bending			Anterior Bending		
	Day 1	Day 2	p-Value	Day 1	Day 2	p-Value
SSL	13.62 ± 7.88	12.32 ± 2.73	0.63	18.31 ± 8.52	18.06 ± 7.39	0.94
LLUL	14.83 ± 6.63	13.34 ± 3.87	0.54	11.93 ± 7.56	11.68 ± 6.47	0.94
ULLT	12.08 ± 8.46	10.70 ± 4.29	0.65	10.53 ± 10.16	8.31 ± 3.28	0.52
LTMLT	8.73 ± 2.29	7.27 ± 2.57	0.20	6.50 ± 2.62	5.00 ± 1.55	0.14
MLTMUT	4.95 ± 1.47	5.81 ± 1.87	0.27	4.63 ± 1.01	4.73 ± 1.02	0.82
MUTUT	5.27 ± 1.82	5.10 ± 1.61	0.82	6.69 ± 3.16	6.26 ± 2.77	0.75
SUT	41.97 ± 14.95	43.61 ± 17.60	0.83	33.67 ± 16.89	35.20 ± 17.04	0.84
SC7	32.98 ± 8.98	34.13 ± 8.59	0.77	27.77 ± 9.98	29.82 ± 12.79	0.69

Similarities of joint angle waveforms were also demonstrated for data collected within and between testing days. CMC values ranged from 0.592 to 0.848 for angles examined within the same day, while between-day CMC values ranged from 0.278 to 0.558 (Table 2). ICC values for the flexion bending task ranged from 0.018 for ULLT angle to 0.713 for the LLUL angle (Table 3). As for lateral bending, ICC values ranged from 0.007 for the ULLT angle to 0.647 for the LLUL angle. The flexion bending task ICC values for the angle calculated using the large spine segments (SUT & SC7) were found to have values of 0.598 and 0.607 respectively, while the large segment lateral bending task ICC values were 0.828 (SUT) and 0.728 (SC7).

Table 2: CMC values for anterior and lateral bending conditions

Angle^0	Lateral Bending		Flexion bending	
	Within	Between	Within	Between
SLL	0.82	0.52	0.77	0.49
LLUL	0.86	0.56	0.79	0.49
ULLT	0.79	0.48	0.72	0.44
LTMLT	0.78	0.51	0.6	0.35
MLTMUT	0.63	0.38	0.48	0.28
MUTUT	0.62	0.38	0.59	0.36
SUT	0.94	0.63	0.82	0.53
SC7	0.95	0.63	0.85	0.56

Table 3: ICC values for the two seated task conditions for eight calculated angles.

Angle^0	Seated Bending	
	Anterior	Lateral
SLL	0.476	0.400
LLUL	0.713	0.647
ULLT	0.018	0.007
LTMLT	0.598	0.196
MLTMUT	0.297	0.111
MUTUT	0.408	0.396
SUT	0.598	0.828
SC7	0.607	0.728

Level Walking

Sagittal plane joint angle patterns during level walking for the angle between the entire back segment and the sacrum (SC7) as well as angles between three adjacent spine segments: SLL, LTMLT, and ULLT are illustrated in Figure 3 for representative subject. Similar motion patterns were found between SC7 and SLL. The segmented angles, SLL, LTMLT and ULLT all were found to have different motions patterns. This suggests that segmented observation of spine motion will capture spinal motions that a large segment back definition will miss in the sagittal plane. All eight spinal angles were shown to have similar between-day ROMs ($p \geq 0.11$; Table 4). CMC values of joint angles defined by the finer segmentation segmented spine angle ranged from 0.62 (UMTUP) to 0.86 (LLUL) for within-day examination and 0.38 (UMTUT) to 0.56 (LLUL) for between–day assessment (Table 5). CMC values of the joint angles defined using large spine segments were .94 (within-day) and .63 (between-day) for SUT and .95 (within-day) and .63 (between-day) for SC7 (Table 5). The ICC values for the sagittal plane segmented angles ranged from 0.051 for the UMTUT and 0.903 for the LLUL (Table 6). While ICC values of the joint angles defined using large spine segments were 0.320 for the SUT and 0.056 for the SC7 (Table 6).

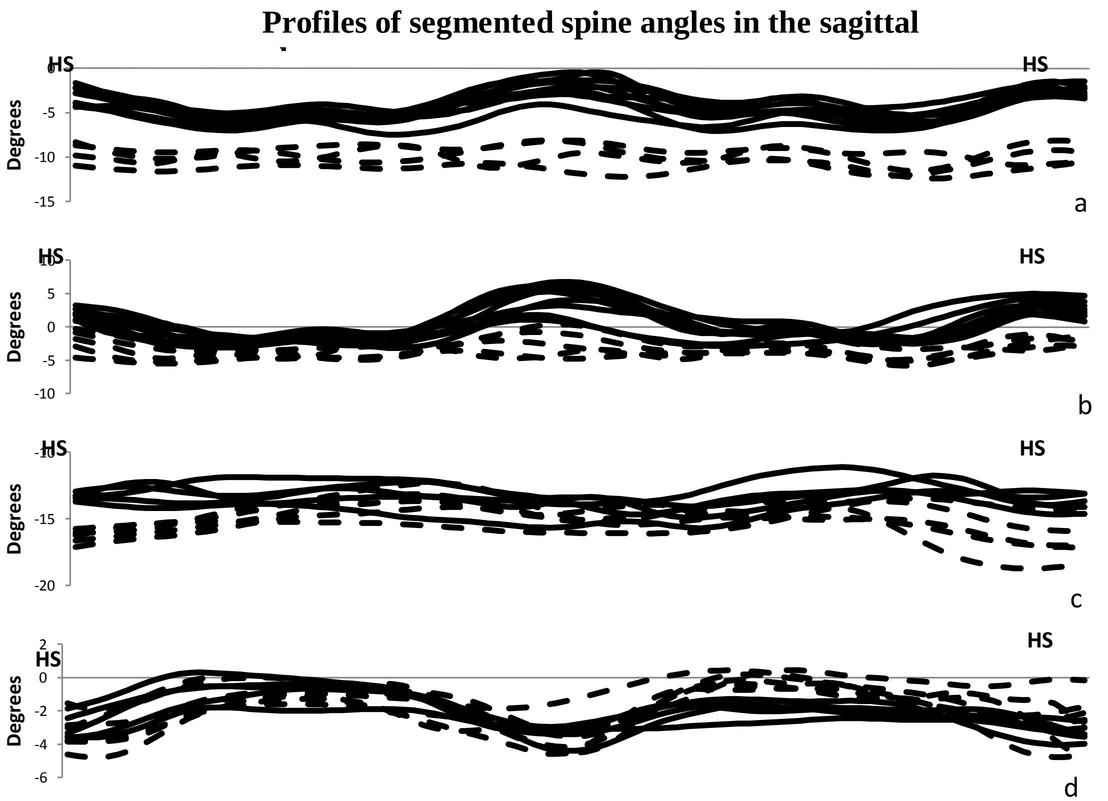


Figure 3. Sagittal plane angles (negative; flexion) of adjacent segments: a)Sacrum to C7, b)Sacrum to LowerLumbar c)LowerThorax to MiddleLowerThorax, d)Upper Lumbar to Lower Thorax from HS to HS for one subject two different testing days. Solid line denotes first day and dashed is second day.

Table 4. ROM values and the respective p-values, in three planes, during walking for all eight calculated angles.

Angle^0	Walking Flexion			Walking Rotation			Walking Lateral Bending		
	Day 1	Day 2	p-Value	Day 1	Day 2	p-Value	Day 1	Day 2	p-Value
SSL	5.37 ± 2.03	5.59 ± 2.32	0.82	13.90 ± 6.82	8.73 ± 2.46	0.04	9.611 ± 3.79	9.80 ± 4.92	0.92
LLUL	5.06 ± 3.31	4.00 ± 2.83	0.45	12.28 ± 6.92	9.96 ± 4.65	0.39	7.13 ± 2.30	7.02 ± 2.68	0.92
ULLT	3.25 ± 1.61	2.63 ± 1.49	0.38	4.93 ± 2.07	4.95 ± 2.14	0.99	3.81 ± 1.77	3.37 ± 2.47	0.65
LTMLT	2.42 ± 0.90	1.83 ± 0.61	0.11	4.25 ± 1.51	3.26 ± 1.07	0.11	2.52 ± 1.22	2.30 ± 0.99	0.67
MLTMUT	6.32 ± 10.28	2.86 ± 2.86	0.33	6.79 ± 8.40	4.29 ± 2.41	0.39	3.76 ± 1.13	2.73 ± 0.68	0.03
MUTUT	4.47 ± 4.56	4.42 ± 5.30	0.98	5.54 ± 7.17	4.93 ± 4.60	0.82	5.39 ± 3.12	4.18 ± 1.19	0.28
SUT	8.25 ± 12.03	7.58 ± 6.39	0.88	17.79 ± 5.12	16.52 ± 6.16	0.62	13.98 ± 5.94	13.46 ± 6.73	0.86
SC7	3.81 ± 1.18	6.46 ± 7.17	0.28	19.25 ± 6.39	19.13 ± 9.37	0.97	7.07 ± 2.06	7.02 ± 1.77	0.96

Frontal plane joint angle patterns during level walking for the angle between the entire back SC7 as well as angles between three adjacent spine segments: SLL, LTMLT, and ULLT were illustrated in Figure 4. Segmented joint angles (SLL, LTMLT and ULLT) were found to have different motion pattern when compared to each other as well as the large segment definition (SC7). SLL and ULLT did show somewhat similar patterns, however, this suggests that segmented observation of spine motion will capture finer spine motions that a large segment back definition will miss in the frontal plane. Seven of eight spinal angles were shown to have similar between-day ROMs except for the LMTUMT angle ($p \leq 0.05$; Table 4). CMC values of joint angles defined by the finer segmentation spine angle ranged from 0.52 (LTLMT) to 0.82 (SLL) for within-days examination and 0.33 (LTLMT) to 0.53 (SLL) for between-days assessment (Table 5). CMC values of the joint angles defined using large spine segments were .86 (within-day) and .60 (between-day) for SUT and .79 (within-day) and .50 (between-day) for SC7 (Table 5). The ICC values for the frontal plane segmented angles ranged from 0.000 for the LMTUMT angle and 0.733 for the LTLMT angle (Table 6). While ICC values of the joint angles defined using large spine segments were 0.862 for the SUT and 0.608 for the SC7 (Table 6).

Transverse plane motion patterns during level walking for the angle between the entire back SC7 as well as angles between three adjacent spine segments: SLL, LTMLT, and ULLT are illustrated in Figure 5. Segmented joint angles (SLL, LTMLT and ULLT) were found to have different motion pattern when compared to each other as well as the large segment discussion (SC7). SC7 and SLL (Figure 5a & 5b) show opposite motion patterns for axial rotation. This suggests that segmented observation of spine motion will

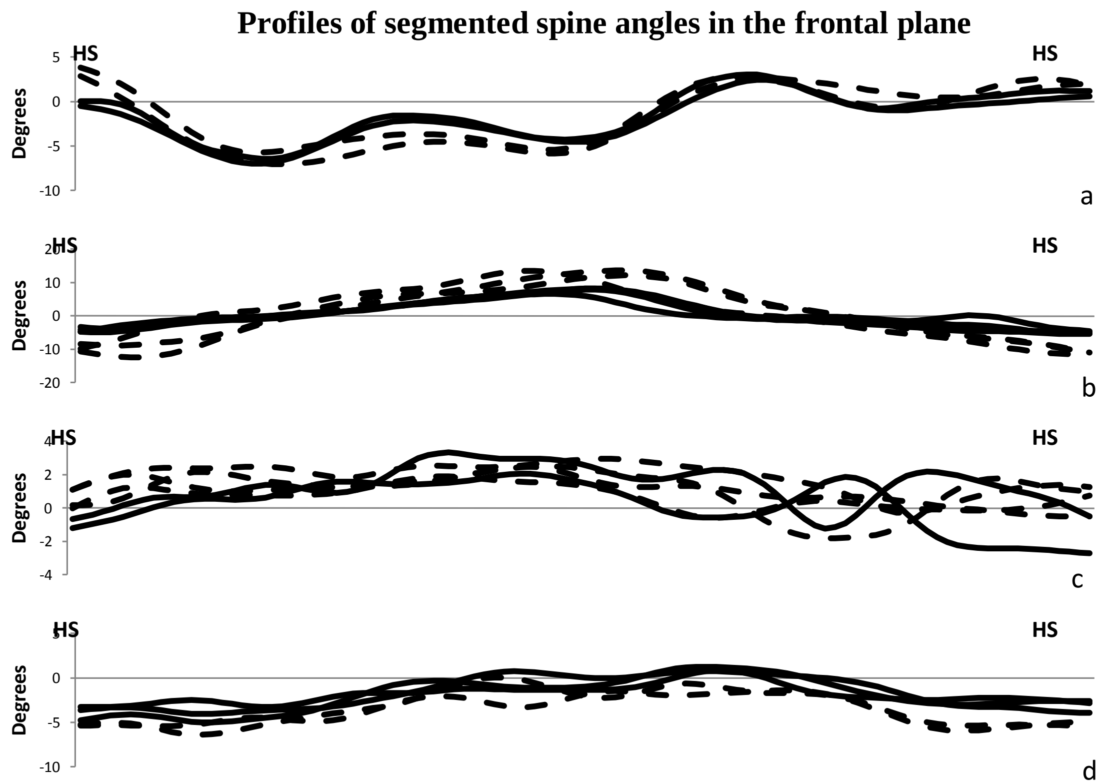


Figure 4. Frontal plane angles (negative flexion) of adjacent segments: a)Sacrum to C7, b)Sacrum to LowerLumbar c)LowerThorax to MiddleLowerThorax, d)Upper Lumbar to Lower Thorax from HS to TO for one subject two different testing days. Solid line is first day and dashed line denotes second day.

record spinal motions that a large segment back definition will miss in the transverse plane of motion. Seven of eight spinal angles were shown to have similar between-day ROMs except for the LMTUMT ($p < 0.05$) angle (Table 4). CMC values of joint angles defined by the finer segmentation spine angle ranged from 0.79 (LLUL) to 0.82 (LMTUMT) for within-days examination and 0.28 (LMTUMT) to 0.53 (SLL & LLUL) for between-days assessment (Table 5). CMC values of the joint angles defined using large spine segments were 0.82 (within-day) and 0.53 (between-day) for SUT and .85 (within-day) and 0.56 (between-day) for SC7 (Table 5). The ICC values for the transverse plane segmented angles ranged from 0.134 for the sacrum to lower lumbar angle and 0.908 for the upper lumbar to lower thorax angle (Table 6). While ICC values of the joint angles defined using large spine segments were 0.680 for the SUT and 0.658 for the SC7 (Table 6).

Table 5: Between and within day CMC values for three planes of walking conditions for eight calculated angles.

Angles^0	Walking Flexion		Walking Rotation		Walking Lateral Bending	
	Within	Between	Within	Between	Within	Between
SSL	0.82	0.52	0.77	0.49	0.82	0.53
LLUL	0.86	0.56	0.79	0.49	0.67	0.44
ULLT	0.79	0.48	0.72	0.44	0.56	0.36
LTMLT	0.78	0.51	0.6	0.35	0.52	0.33
MLTMUT	0.63	0.38	0.48	0.28	0.65	0.44
MUTUT	0.62	0.38	0.59	0.36	0.68	0.46
SUT	0.94	0.63	0.82	0.53	0.86	0.60
SC7	0.95	0.63	0.85	0.56	0.79	0.50

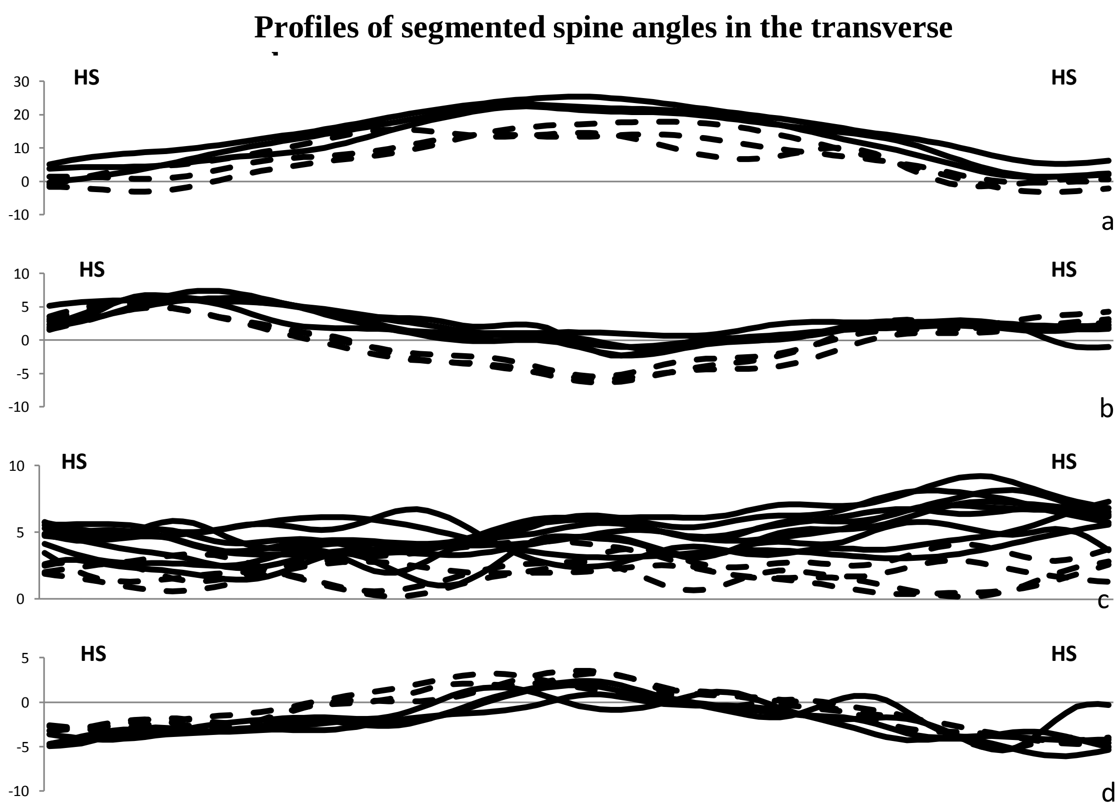


Figure 5. Transverse plane angles (negative; flexion) of adjacent segments: a)Sacrum to C7, b)Sacrum to LowerLumbar c)LowerThorax to MiddleLowerThorax, d)Upper Lumbar to Lower Thorax from HS to TO for one subject two different testing days. Solid line is first day and dashed line is second day.

Table 6: ICC values for the walking condition of eight calculated angles.

Angle^0	Walking		
	Flexion	Rotation	Frontal
SLL	0.495	0.134	0.465
LLUL	0.903	0.244	0.489
ULLT	0.848	0.908	0.713
LTMLT	0.059	0.611	0.733
MLTMUT	0.471	0.463	0.000
MUTUT	0.051	0.259	0.109
SUT	0.320	0.680	0.862
SC7	0.056	0.658	0.608

Discussion

The purpose of this study was to assess the validity of an in-vivo motion capture method which would demonstrate the need of quantifying individual spinal movement during locomotion. Qualitative observations of the time histories revealed different motion patterns for different spinal segments and the corresponding adjoining angles when compared to the ‘entire’ back (single) segment during gait (Figures 3 - 5). In sagittal plane motion (Figure 3), there were similarities between SC7 and SLL. This could be expected because, although the definitions of the proximal segments were different, the measured joint angle (relative position between sacrum sand lower lumbar segments) is the same. The magnitude of the measured motion is much less in the segmented back marker set (SLL) than the entire back definition (SC7). It could also be clearly observed that distinct differences in motion patterns were displayed by angles derived from more superior paired segments (Figure 4; LTMLT & ULLT). Furthermore, unlike sagittal plane motion, the frontal plane angles derived between the sacrum and the entire back segment (Figure 4a, SC7) and between the SLL (Figure 4b) were different. This suggests the size of the superior segment changes the motion pattern in the frontal plane. SLL and ULLT did show somewhat similar patterns, indicating the lumbar area of the spine moves in a similar way during gait (Figure 4). Segmented spine motion (Figure 5, b-d) in the transverse was found to have different patterns than the entire back motion (Figure 5a). SC7 and SLL (Figure 5a & 5b) show opposite motion patterns for axial rotation. This suggests that the segmented approach to quantifying spine motion is able to record motions that the large segment (SC7) cannot. The segmented approach SLL is recording the motion between the sacrum and the lower lumbar segment, while the large segment approach (SC7) is recording the rotation of the shoulders with respect to the sacrum. This is due to how this segment is defined superiorly using the left and right shoulder markers. Taken together, these findings suggested that different spinal segments exhibit different movement patterns during gait, and the spinal motion quantified using a single trunk (or back) segment could not reveal unique movement features associated with different areas of the spine.

Range of motion values of most joint angles were found to be repeatable between testing days during gait (Table 4). Possible explanations for the non-repeatable ROMs could be a fatigue muscle response (specifically superior erector spine group) between different days. Subjects were not instructed to restrict their physical activities prior to testing. In addition, the testing time in the day was not controlled for the subject. Therefore, if a subject was tested at different times in the day (e.g., early morning vs. later afternoon), there is a possibility that fatigue may have contributed to different spinal motions. Additionally, marker placement could be a source of error which would cause the ROM to be non-repeatable between testing days. Precautions were taken to address this possible limitation by having only one researcher place markers on all the subjects.

The overall average of the within-day CMC values were higher than those calculated for between-day evaluations, which agrees with previous studies (Ferrari et al., 2010; Kadaba et al., 1990). It has been suggested that a CMC value larger than 0.9 are considered strong correlations, 0.5 is considered to be moderate, and less than 0.25 is considered to have a weak correlation (Kadaba et al., 1990). Overall, the large segments (SC7 & SUT) showed larger CMC values than the segmented spine angles. This is not surprising, as it has been recently reported that the CMC calculation may not be as dependable when determining reliability between waveforms from with small ranges of motion (Ferrari et al., 2010). This is particularly important for the current study due to the small range of motion exhibited between the adjacent spinal segments. Another observable trend was the CMC values generally decrease as the angle of interest is more superior. This is true for both within-day and between-day and among the sagittal and transverse planes. The frontal plane of motion was found to have larger CMC values in the more inferior angles (SLL, LLUL) as well as the superior angles (MLTMUT and MUTUT), Table 5. This may suggest that the transition from the lumbar spine to the thoracic spine and the lower thoracic spine do not produce repeatable segmental motion. SLL, LLUL and UTTL were found to have larger CMC values then the more superior angles in the sagittal and transverse planes of motions, suggesting the more inferior segments produce more repeatable data than the upper segments.

There were wide range distributions of ICC values for the angles between adjacent segments and that defined using a large back segment. Similar to the current study, wide range of ICC values has been previously reported for spatiotemporal parameters (e.g., stride width, stride length and cadence) in children and adults (Olsson et al., 2004). Additionally, large variations in ICC values have been found during treadmill walking (N. Taylor et al., 1996). This erratic range of ICC values suggests that only certain segmental angels resemble each other between days. Unlike CMC, there were not any trends associated with the ICC values. The large back segment ICC values were not consistently higher than the segmented angle ICC values. Furthermore, the ICC values did not decrease from inferior angle to superior segmented angle as was reported in the CMC repeatability measures. According to the ICC segmented spine angle values, LLUL (0.903) and ULLT (0.848) were found to be the most reproducible in the sagittal plane, while ULLT (0.908) was found to have the only reasonable ICC value in the transverse plane of motion (Table 6). In the frontal plane, the ULLT (0.713) and LTMLT (0.733) angles were the most reproducible. Large segment spine definition ICC values ranged from 0.056 (sagittal plane SC7) to 0.862 (frontal plane SUT). This high variably coupled with no observable trend in the ICC values suggests that ICC may not be the most ideal calculation for reproducibility/reliability for a segmented spine. This could be due to the sometimes high standard deviations which can be found in these angles. The large standard deviations in the ensemble average data are not surprising as multiple motion patterns have been reported in the healthy spine (Gatton & Pearcy, 1999). Due to the fact, the ICC calculation depends on the standard deviation; it may not be the ideal statistic for determining reproducibility/reliability for this marker set.

This new marker set and protocol to quantify spine segmental joint angles during ambulation could provide new information regarding the mechanics of the spine. It must be noted that only some of the segmental joint angles should, at this point, be considered. The CMC values for the three inferior segmental angles (SLL, LLUL and ULLT) were the highest for all three planes of motion. It is therefore the current suggestion to focus the investigation of segmented spine angles during ambulation to the three most inferior angles. In part, due to the larger repeatability values and a large majority of back pain does present in the low back. Therefore an improved understanding this region of the spine could provide new biomechanical insights to the mechanisms of back pain.

To our knowledge this is the first study to investigate the reliability and usability of using surface markers to assess multi-segmental spinal motion during over ground walking. Our findings demonstrated the ability to detect different motion patterns between different segmented levels of the spine as well as differences when compared to the large segment quantifications methods of the spine. This new approach in spinal quantification could allow for a better understanding of possible degenerative mechanisms that exists as a result of normal ambulation. The identification of these mechanisms will allow for improved future treatment methods and possibly the development of preventative back pain screenings.

Bridge

Chapter II examined the feasibility of using a newly proposed in-vivo biomechanical marker set and protocol to assess multi-segmental spinal motion during gait. In Chapter III, this newly developed marker set will examine the potential differences in segmental spinal kinematics during multiple ambulatory activities of daily living.

CHAPTER III

SPINE MOTION DURING ACTIVITIES OF DAILY LIVING IN YOUNG ADULTS

This chapter was developed by Li-Shan Chou, Ph.D. & Scott P. Breloff. Dr. Chou contributed substantially to this work participating in the development of methodologies and providing invaluable critiques and substantial editing advice. Scott P. Breloff was the primary contributor to the development of the protocol, data collection, data analysis and did the writing.

Introduction

Various forms of acute back pain can affect up to 65% of the American population (Walker, 2000). This aliment is in the top five reasons for people to miss work, visit a doctor, have surgery or be admitted to the hospital and can be caused by several different mechanisms (NINDS, 2011; V. M. Taylor et al., 1994). It has been reported that individuals who are experiencing some forms of back pain exhibit a lower physical activity level, less standing time, lower step frequency and more sedentary time during the evening (Spenkelink et al., 2002). The frequency of back pain and the lack of treatment methods (Bigos et al., 1994) are the motivation for this current investigation. It is the intent of the current study that the results will be used to better understand biomechanical mechanisms which could lead to back pain, thereby aiding in the development of better treatment methods and new preventative procedures.

It has been reported that back pain can affect gait function (Spenkelink et al., 2002). It is therefore beneficial to understand the mechanics of the back and spine during gait. Many of the methods used to quantify back motion (lifting or gait) either detected the motion using large spinal segments (Chaffin et al., 2006; Crosbie et al., 1997b; Kadaba et al., 1990; Park & Chaffin, 1974) or used bone pins to assess movement of specific spinal segments (Rozumalski et al., 2008). These two approaches either neglected the complexity of multi-segmental anatomy of the spine or were too invasive to be used in individuals with back pain. Therefore a procedure which has the ability to record detailed segmented spine motion while maintaining its usefulness in a clinical setting would be most advantageous in furthering the understanding spine of dynamics during multiple ambulatory tasks.

The term "activities of daily living," or ADLs, refers to the basic tasks of everyday life, such as eating, bathing, dressing, toileting, transferring and ambulation (Wiener, Hanley, Clark, & Van Nostrand, 1990). Healthy individuals move from place to place by walking (gait), sometimes certain external perturbations may be introduced. For example, crossing the street may require an individual to step over a curb, or living in a house with more than one level which requires numerous trips up and down stairs. These ambulatory activities of daily living (level walking, obstacle crossing, stair ascent & descent) are the focus of this investigation and how segmented back motions can be altered by each task.

Obstacle crossing has been examined and was found to perturb the center of mass (COM) motion in the frontal plane (Chou et al., 2001). Although the COM is located in the trunk, it is not a direct measure of the spinal movement. Results from this study may suggest that an obstacle crossing task would induce changes in segmented spine kinematics. Additionally, Hahn and Chou (2003) investigated how obstacle crossing could influence individual segment motion and determine instability in the elderly and showed trunk segment angles would change has a result of the increasing height of the obstacle. The reported trunk segment angle was defined using one large segment (left and right shoulder markers superiorly and a sacral marker inferiorly) and is the motion about the global coordinate system. This information does suggest that trunk motion is influenced by obstacle crossing; however it does not provide information on the vertebral motion.

Range of motion differences have been found in the trunk during stair negotiation in all planes of motion (Krebs et al., 1992). Recently, J. K. Lee and Park (2011) used a magnetic tracking device, placed at the 12th thoracic and 1st sacral spinous process, to measure spinal motion during stair walking. Clear differences were reported in spinal motion between the level walking and staircase walking conditions, particularly in regards to the motion pattern and ROM for flexion/extension and lateral bending of the spine. Direct conclusions could not be drawn to vertebral motion in these studies due to the large segment definitions of the trunk. These results suggest that spine motion may be influenced during stair negotiation. The application of an in-vivo multi-segmented spine marker set could further the knowledge of spinal mechanics during these (stair ascent & descent) ambulatory activities of daily living.

Range of motion of spinal angles has been reported to be different between healthy individuals and patients with back pain (Mayer, Tencer, Kristoferson, & Mooney, 1984; Neblett, Mayer, Brede, & Gatchel, 2010). Therefore, the ROM was selected as a biomechanical measure to determine if differences exist in segmental spinal motion between numerous ambulatory tasks of daily living. Peak joint angle has been reported in the lower extremity during healthy gait and stair negotiation (Perry, 1992; Samuel, Rowe, Hood, & Nicol, 2011; Varin, Lamontagne, Beaulieu, & Beaulé, 2011; D. A. Winter, 2009). Only a few studies reported peak angles in the upper extremity (Tester, Barbeau, Howland, Cantrell, & Behrman, 2012). The instance (timing) at which the peak occurs has previously been used to compare differences in lower extremity kinematics (Jorrakate, Vachalathiti, Vongsirinavarat, & Sasimontonkul, 2011; McLean, Huang, & van den Bogert, 2005). The timing of the upper extremity peak angle has been used previously to coordinate differences between the lower and upper extremity (Tester et al., 2012). Thus, along with the range of motion, it was determined that the peak angle and the timing of these peaks may be affected with the introduction of ambulatory activities of daily living.

The results of these previous studies suggest that obstacle crossing and stair negotiation may influence the motion of the trunk, but do not discuss the motion of the spine specifically. Therefore the inclusion an in-vivo marker set containing different spine segments will provide a more comprehensive understanding to the motion of the spine during ambulatory activities of daily living. A more precise knowledge of the spine motion during ambulation may be able to detect different injury mechanisms which could cause back pain. The understanding of these mechanisms could lead to better treatment protocols and new preventative procedures to help individuals who suffer from back pain. Thus, the purpose of this study was to examine differences in the kinematics of individual spinal segments of young adults when performing different activities of daily living, including level walking, obstacle crossing, and stair ascending/descending. It was hypothesized that different activities of daily living will exhibit different kinematics at different spinal segments.

Methods

Subjects

Fourteen healthy young adults (7 males/7 females; mean age: 27.9±5.9 years, mean height: 176.0±27.7 cm, mean mass: 67.8±17.2 kg) were recruited from the university community to participate in the study. Subjects had no history or clinical evidence of neurological, musculoskeletal or other medical conditions affecting gait performance, such as stroke, head trauma, neurological disease (i.e. Parkinson's, diabetic neuropathy), visual impairment (not correctable by lenses) and dementia. All subjects reviewed and signed an informed consent approved by the Institutional Review Board.

Experimental Protocol

Subjects were asked to wear spandex shorts with no shirt for men and with dance leotard with open back for women. They subsequently performed four different tasks while with bare feet: level ground walking (W), obstacle crossing (OC), stair ascent (SA) and stair descent (SD). The task order was randomly presented for each subject. The level walking task required subjects to walk along a 10 meter long walkway. For obstacle crossing task, subjects were asked to initiate walking from a distance which allowed at least 3 steps prior to encountering the obstacle, step over the obstacle, and continue walking. The obstacle was set at 10% of body height and made of a polyvinyl chloride (PVC) pipe measuring 1.5 m long and a diameter of 2.5 cm, which was presented to the subjects prior to obstacle crossing trials (Hahn & Chou, 2004). During the SA, subjects were asked to approach the stairs while walking on level ground, ascend the stairs, and continue walking to the end of the elevated walkway. The starting position for each subject was adjusted to allow at least three steps before stepping onto the first stair.

Subjects initiated their SD trials from the back end of the elevated walkway, descended the stairs, and continued walking for several steps.

Whole body motion analysis was performed with a ten-camera motion analysis system (Motion Analysis Corporation, Santa Rosa, CA). Sixty-two retro-reflective markers (diameter=14mm) were placed on the subject. In addition to a whole body marker set (Hahn and Chou, 2004), eight markers were placed directly on the palpated spinous processes of the following vertebrae; Cervical 7 (C7), Thoracic 3 (T3), Thoracic 6 (T6), Thoracic 9 (T9), Thoracic 12 (T12), Lumbar 3 (L3), Sacrum 1 (S1) & Sacrum 5 (S5). Two makers were placed on the left and right posterior superior iliac spine (PSIS), and the remaining markers were placed 50mm to the left and right of the spinous process markers, except for S1 and S5 (Figure 1). Three-dimensional marker position data were collected at 60 Hz and low-pass filtered using a 4^{th} order Butterworth filter with the cutoff frequency set at 5 Hz.

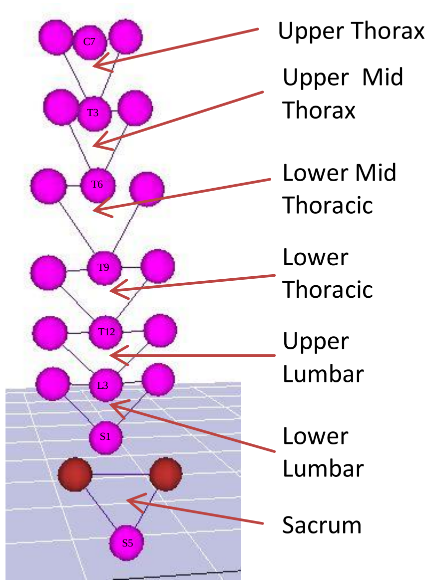


Figure 1. Segmental spine maker set with all six adjacent segments which angles were calculated. Only the three most inferior three (SLL, LLUL and ULLT) were analyzed.

Identical force plate configurations were used for the non stair related tasks [Walking (W) & Obstacle Crossing (OC)]. Three force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were placed in series and embedded level into the laboratory floor. The first two force plates were immediately adjacent to one another, and the third plate was separated by a distance of 15cm. This setup was to accommodate subjects walking with different step lengths. Gait events such as heel strike (HS) and toe off (TO) were detected using the vertical ground reaction force (GRFv). Heel strike was determined to occur when the GRFv was greater than 10% of the maximum GRFv, and toe off was determined to occur when the GRFv was less than 10% of the maximum GRFv (Ghoussayni et al., 2004; Hreljac & Marshall, 2000; Mickelborough et al., 2000).

For stair ascending (SA) and descending (SD) tasks, a staircase including three steps was used (Figure 2). Each step had a rise of 17.8 cm, a run of 30.5 cm and a width of 80 cm, forming a stair angle (rise/run) of 30^0 (H. J. Lee & Chou, 2006) A total of four force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were used to obtain ground reaction force data during SA and SD trials. Two force plates were embedded level into the laboratory floor and two made up the steps (Figure 2).

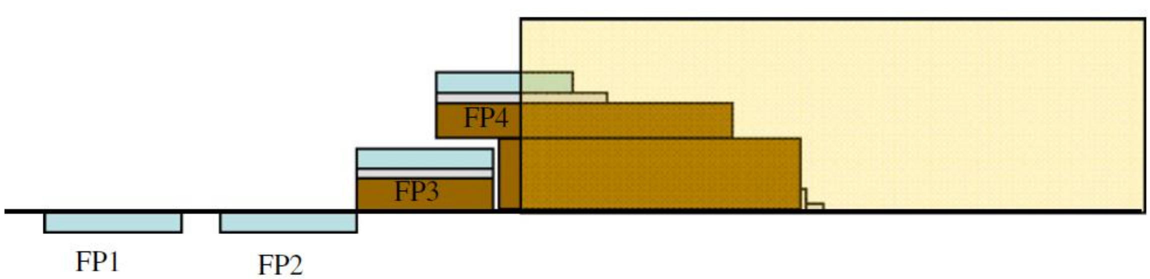


Figure 2. Depiction of stair set up with embedded force plates. Reprinted with permission from (H. J. Lee & Chou, 2006).

Marker position data were analyzed for one activity cycle of each condition. Five trials were captured for each condition. A gait cycle during level ground walking was defined as the time interval between two consecutive ipsilateral heel strikes (FP 3 to FP 1; Figure 3a). The obstacle crossing stride was defined as the heel-strike of the leading limb before the obstacle to the heel-strike of the same limb after clearing the obstacle (Figure 3b). Stair ascent was examined for the duration between consecutive ipsilateral heel strikes of last level ground contact and the second stair (FP 2 to FP 4; Figure 3c), and stair descent was examined consecutive ipsilateral heel strikes following first step down to ground (FP 4 to FP 2; Figure 3d)

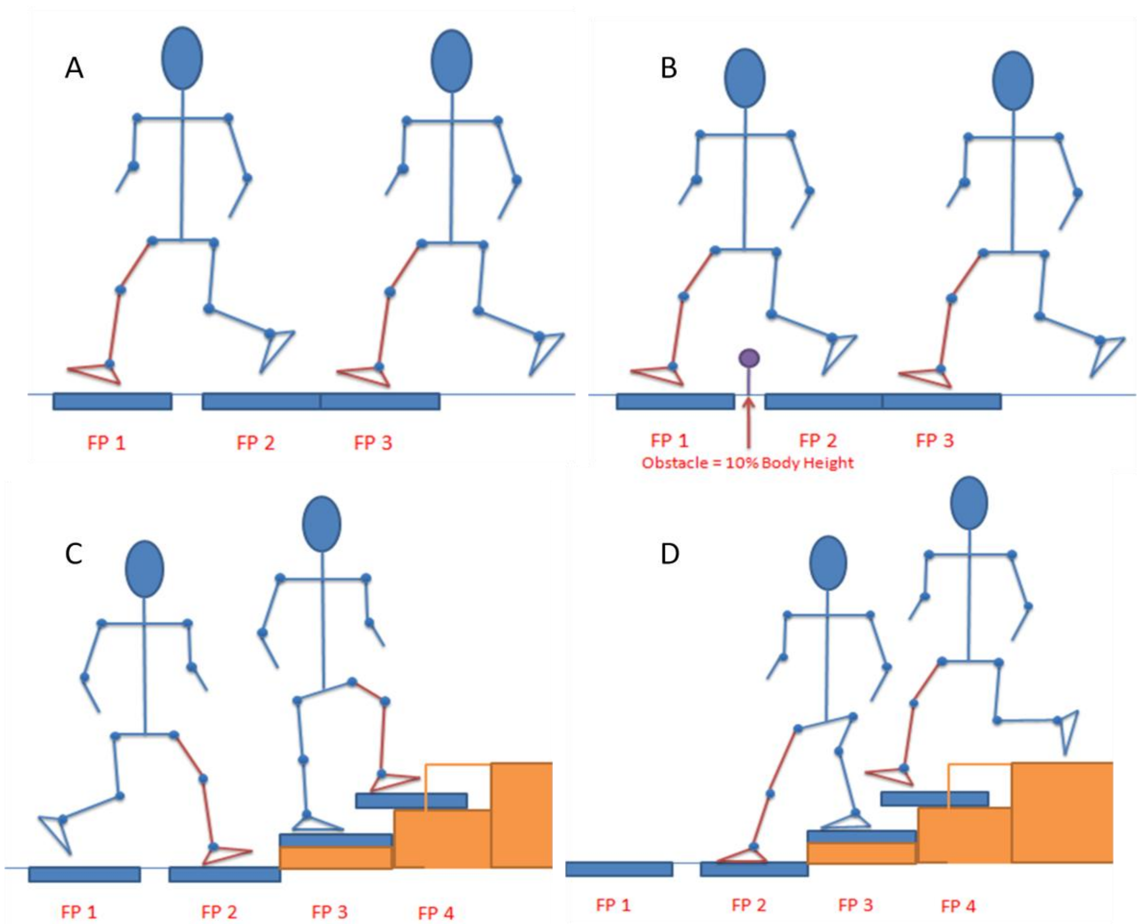


Figure 3. Definition of each task. (A) Level walking (W) – ipsilateral heel strikes, (B) Obstacle Crossing (OC) – Leading limb ipsilateral heel strikes, (C) Stair Ascent (SA) - ipsilateral heel strikes, (D) Stair Descent (SD) - ipsilateral heel strikes.

Data Analysis

A MATLAB® (Mathworks, Natick, MA) program was developed to calculate six three dimensional adjacent segmental spinal angles. For this study, it was decided to focus on the three most inferior adjacent spine segments due to the fact that most back pain presents in the low back (NINDS, 2011). The segments in the current study are labeled as: sacrum to lower lumbar [SLL], lower lumbar to upper lumbar [LLUL] and upper lumbar to lower thorax [ULLT] (Figure 1).

Peak angles, timing of the peak angle and range of motion have been used previously to describe spinal motion during ambulatory activities of daily living (Jorrakate et al., 2011; Mayer et al., 1984; McLean et al., 2005; Neblett et al., 2010; Perry, 1992; Samuel et al., 2011; Varin et al., 2011; D. A. Winter, 2009). In the current study, these two biomechanical parameters will be examined at different spine levels and during tasks of daily living. Peak angles will be examined in three different planes of motion. In the sagittal plane, the peak flexion angle is presented. Spine flexion is considered forward/flexion bending of the spine. In the frontal plane, peak ipsilateral bending is measured. This definition indicates the spine is bending toward the initial contact leg, or bending to the ipsilateral side. In the transverse plane, peak contralateral axial rotation is discussed. After initial heel strike, the spine bends away from the striking leg or toward the contralateral leg. The timing of the peak angle (index) is the occurrence of the peak angle during the task. In the current study, this biomechanical parameter is used to describe the gait event that is occurring during the peak angle in each plane of motion. Range of motion is the total excursion of the segmental joint angle during each task.

Data for two of the three biomechanical outcome measures were analyzed using a two-way within factor analysis of variance. The dependent variables were peak flexion angle (deg) and range of motion (deg). Task was a within subject effect with four levels: (a) level walking, (b) obstacle crossing, (c) stair ascent, and (d) stair descent. The second factor, also a within subject effect, was spine level with three levels: (a) Sacrum to Lower Lumbar (SLL), (b) Lower Lumbar to Upper Lumbar (LLUL) and (c) Upper Lumbar to Lower Thorax (ULLT). For all outcome measures, except for maximum peak angle in the sagittal plane of motion and the range of motion in the sagittal plane, adjusted p-values (Grennhouse-Geisser) were used to evaluate within subject effects because the assumption of sphericity was evaluated with the Mauchly Sphericity Test and found to be non-tenable, $p < 0.05$. If a significant interaction between factors was detected, pair-wise comparisons were applied to identify the differences. However, if the interaction was not significant then the main effects of each factor was discussed. The statistical software PASW (version 18, IBM., New York, NY) was used for all statistical analyses. The level of significance for these statistical tests was set at 0.05. Additionally, partial eta squared ($\eta 2$) were calculated as effect sizes (ES) for all variables to assist in the explanation of any trends. An ES was considered small (0.01), medium (0.06), and large (0.14) respectively (Cohen, 1988). This is interpretation of the eta squared statistic is appropriate because it is used to as an index of the strength of association between an independent variable (spine level and task) and a dependent variable (ROM and peak angles) that excludes variance produced by other factors (Pierce, Block, & Aguinis, 2004).

Results

Figure 4 presents ensemble average segmental spine joint angles at SLL, LLUL and ULLT. Initial heel strike (HS) is at 0% stride and ipsilateral HS is at 100% stride. The largest flexion angles occur at the SLL angle during stair ascent. The total excursions in the sagittal plane decease as the joint becomes more superior. Compared to the sagittal plane, the ipsilateral bending angles have much smaller angles. Contralateral axial rotations have similar patterns for all conditions. The largest excursion occurred in the sagittal plane at the SLL joint, and the total contralateral axial rotation excursion decreases as the joints are location more superiorly in the spine.

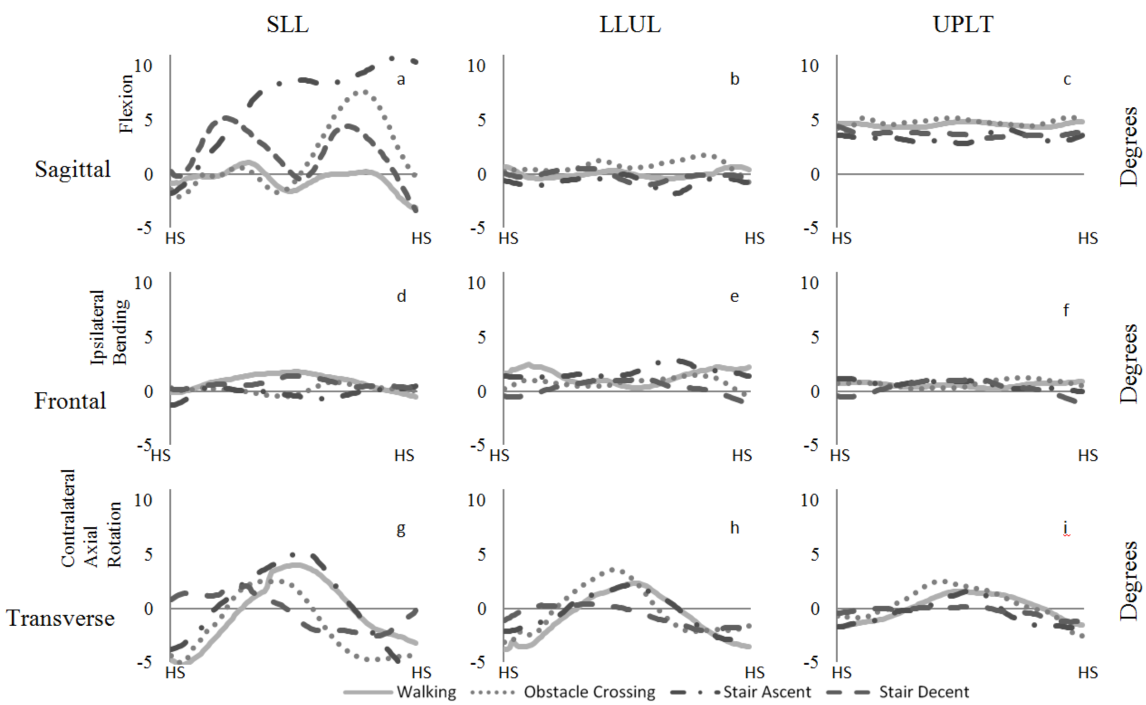


Figure 4. Ensemble average segmented spinal angles; (a) sacrum to lower lumbar sagittal, (b) lower lumbar to upper lumbar sagittal, (c) upper lumbar to lower thorax sagittal, (d) sacrum to lower lumbar frontal, (e) lower lumbar to upper lumbar frontal, (f) upper lumbar to lower thorax frontal, (g) sacrum to lower lumbar transverse, (h) lower lumbar to upper lumbar transverse, (i) upper lumbar to lower thorax transverse. Positive values in the sagittal plane indicate flexion; in the frontal plane, ipsilateral bending is positive. In the transverse plane, contralateral axial rotation is positive.

Sagittal Plane Motion

There was not a significant interaction between spine level and task in the sagittal plane of motion on the peak flexion angle (p = 0.426). There was a significant main effect of spine level on peak flexion angles (Table 1). Conversely, pairwise comparisons of the marginal means of spine levels values were not significant. A significant main effect of the marginal means of task on peak flexion angle was found. Pairwise comparisons of task marginal mean values revealed SA ($6.93^0 \pm 1.17^0$) and OC ($5.91^0 \pm 1.24^0$) had significantly larger angles than SD ($3.95^0 \pm 0.93^0$). Furthermore, additional pairwise comparisons between cells of all spine levels and all tasks were not significant (Figure 5).

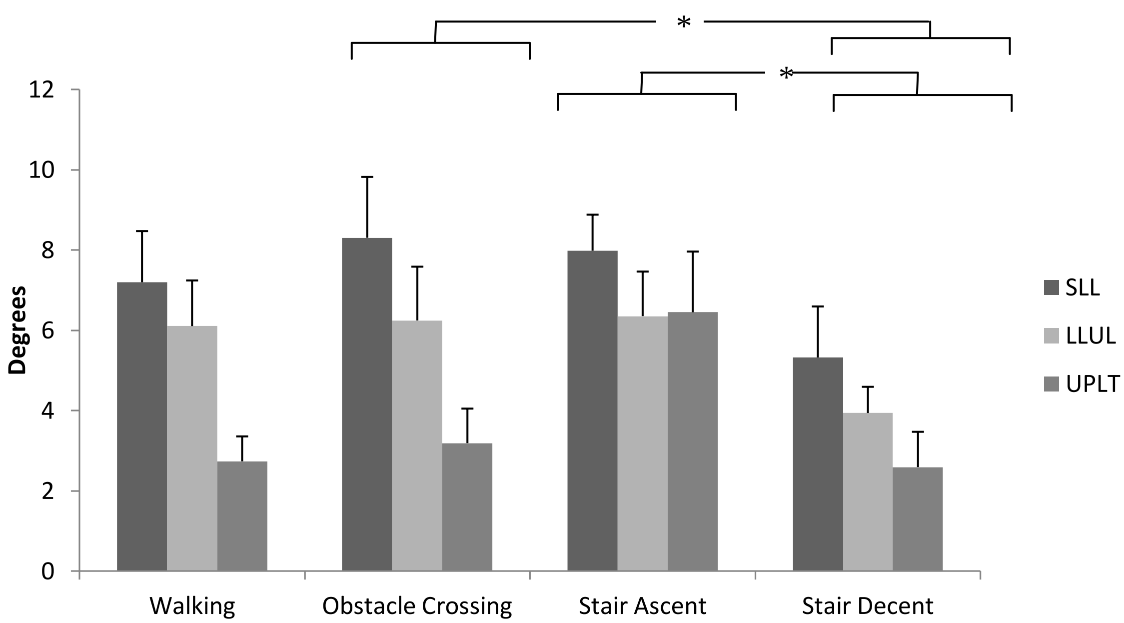


Figure 5. Significant differences between spinal levels with each condition for peak flexion angles in the sagittal plane. *indicates statistically significant change.

The timing of the peak flexion angle was used to determine what gait events are occurring when segmental spine peak flexion is produced (Table 1). The timing of the peak flexion angle (index) is the period of time during the defined gait cycle when the peak angle was observed in a percent. For the W task, SLL angle occurred just before heel strike (HS) of the contralateral limb (~39% gait cycle [GC]). The LLUL peak flexion angle was observed during HS of the contralateral limb. Peak flexion at ULLT angle was shortly after contralateral HS as the heel rocker was engaged. Obstacle crossing presented with different gait events during peak flexion angles. The SLL angle occurred just after the ipsilateral toe off (~63% GC). Peak flexion angle presented in the LLUL angle just before contralateral HS, while the peak flexion angle for ULLT occurred just before the toe was to cross the obstacle. All three segmental peak spine flexion angles during SA presented during contralateral heel strike on the first stair (~45-49% GC). During SD, all the peak flexion angles presented as the contralateral foot was approximately in initial-swing (Figure 6).

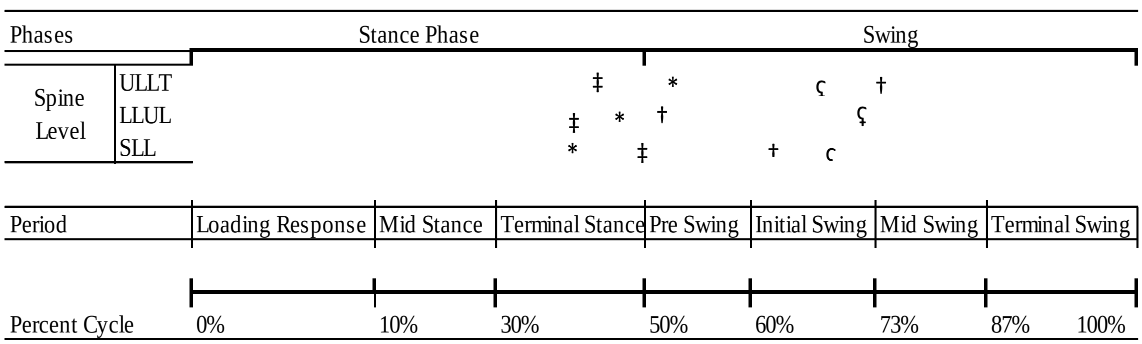


Figure 6. Where peak angles occurred in sagittal plane during gait. * Denotes walking condition; † obstacle crossing; ‡ stair ascent; ʕ stair descent

There was not a significant interaction between spine level and task in the sagittal plane range of motion [ROM] (p = 0.255). There was a significant main effect of spine level for sagittal plane range of motion (Table 1). Similar to the peak angles, the SLL joint exhibited the greatest range of motion across all of the tasks. Post hoc pairwise comparisons of the spine level marginal means revealed SLL ($13.27^0 \pm 1.66^0$) was significantly larger than ULLT ($7.34^0 \pm 0.83^0$) and LLUL ($10.51^0 \pm 1.63^0$) was significantly larger than ULLT (Figure 7a). There was a significant main effect of task on sagittal plane range of motion. Follow up pairwise comparisons on the marginal means of task showed W ($9.74^0 \pm 1.28^0$) to be significantly smaller than SA ($14.26^0 \pm 1.60^0$). Additionally, SA was significantly smaller than SD ($6.97^0 \pm 0.86^0$) [Figure 7b]. Further post hoc pairwise comparisons of all task and all spine level cells were not significant.

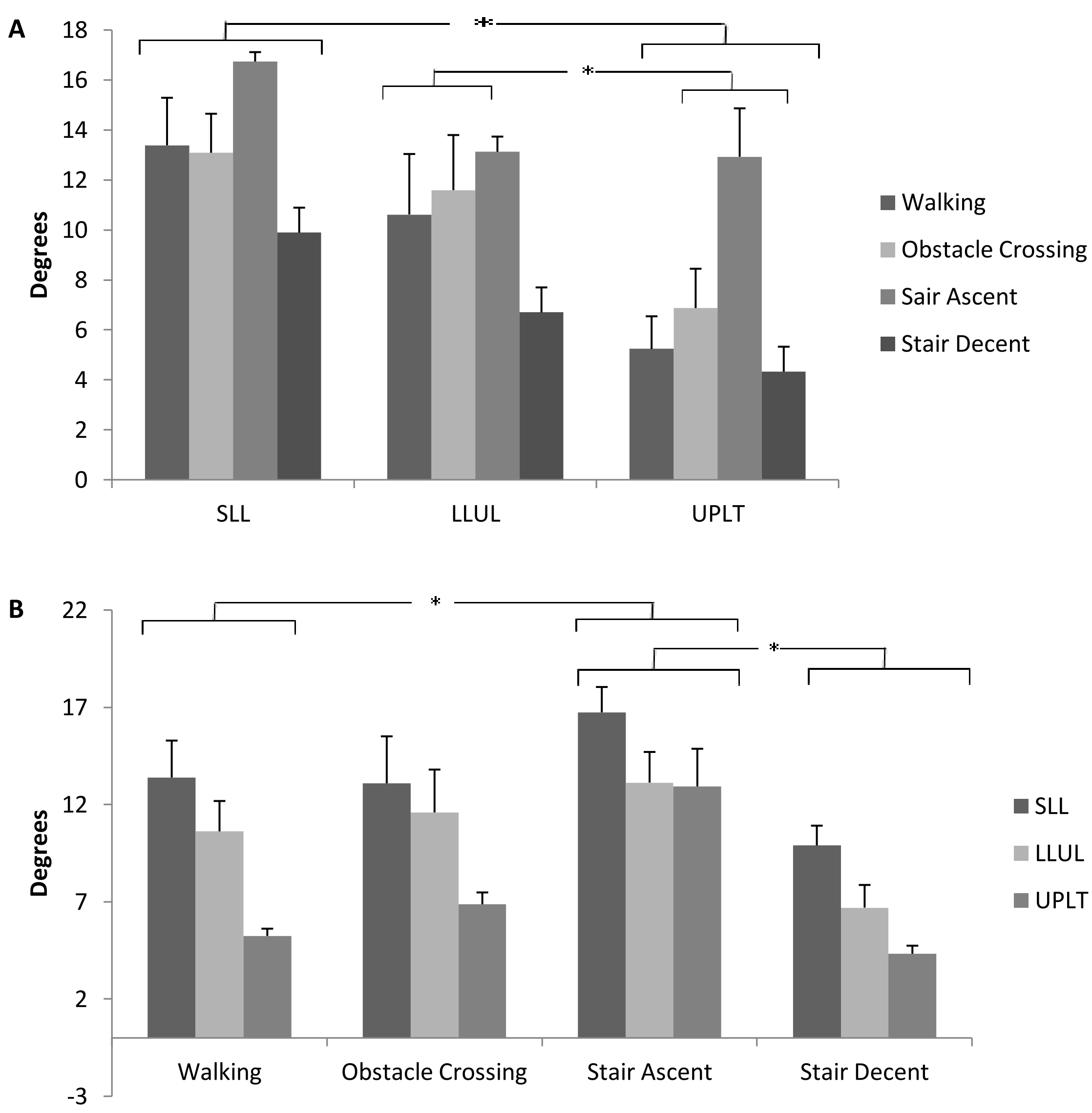


Figure 7. Range of motion in the sagittal plane of motion for various activities of daily living. A) is spine level ROM for each task of daily living. B) is the task specific ROM for all spine levels.

Table 1. Flexion outcome measures of maximum peak angle, maximum peak angle index and range of motion.

	SLL	LLUL	UPLT	p-value
Maximum Peak Angle (deg) Mean (SEM)				
Spine Level (Overall)				.006
Task (Overall)				.005
Walking	7.20 ± 1.27	6.10 ± 1.14	2.73 ± 0.62	
Obstacle Crossing	8.30 ± 1.52	6.24 ± 1.34	3.19 ± 0.86	
Sair Ascent	7.98 ± 0.90	6.35 ± 1.11	6.45 ± 1.51	
Stair Decent	5.32 ± 1.27	3.94 ± 0.65	2.59 ± 0.88	
Maximum Peak Angle Index(%) Mean (SEM)				
Spine Level (Overall)				.098
Task (Overall)				<.001
Walking	39.41 ± 6.91	48.54 ± 6.59	56.44 ± 5.81	
Obstacle Crossing	63.10 ± 6.63	44.89 ± 5.84	73.25 ± 4.87	
Sair Ascent	48.84 ± 2.85	45.32 ± 3.40	47.14 ± 4.43	
Stair Decent	65.78 ± 3.66	70.21 ± 3.80	64.86 ± 5.28	
Range of Motion (deg) Mean (SEM)				
Spine Level (Overall)				.001
Task (Overall)				<0.001
Walking	13.38 ± 1.90	10.62 ± 1.56	5.24 ± 0.38	
Obstacle Crossing	13.09 ± 2.42	11.59 ± 2.20	6.87 ± 0.61	
Sair Ascent	16.74 ± 1.30	13.12 ± 1.58	12.93 ± 1.94	
Stair Decent	9.89 ± 1.02	6.70 ± 1.16	4.33 ± 0.41	

p: main effect significance level.

Frontal Plane Motion

There was not a significant interaction between spine level and task for the maximum peak angle in the frontal plane of motion (p = 0.169). Additionally, main effects for spine level and task were not significant for peak ipsilateral bending angles (Table 2; Figure 8).

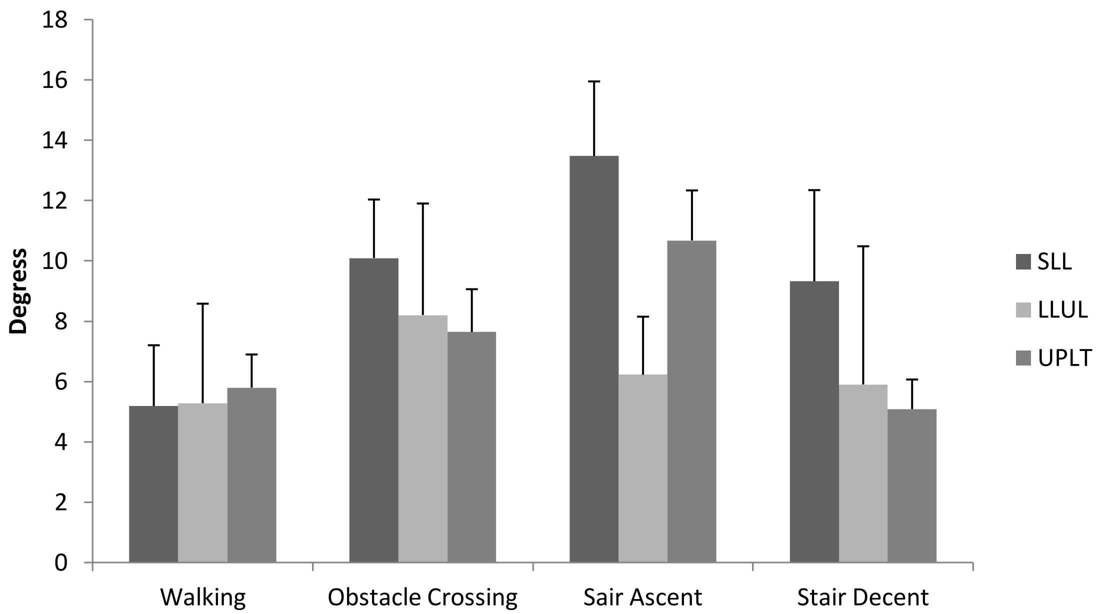


Figure 8. Peak ipsilateral bending angle for various activities of daily living.

During walking, the peak ipsilateral bending angles for SLL, LLUL and ULLT were observed during the action of the weight acceptance period the contralateral leg shortly after HS (~56% GC). Obstacle crossing produced the peak ipsilateral bending angles for SLL, LLUL and ULLT during terminal stance and initial swing of the ipsilateral limb (~70% GC). The SA task, produced peak ipsilateral bending for the SLL angle during initial and mid swing of the ipsilateral leg while the ipsilateral limb was just posterior of the contralateral limb (~74% GC). The LLUL and ULLT joints had peak ipsilateral bending angles occur during the action of the weight acceptance period the contralateral leg shortly after HS. Stair descent produced peak ipsilateral bending angles for the SLL, LLUL and ULLT joints during late swing of the contralateral limb (~47% GC) [Figure 9].

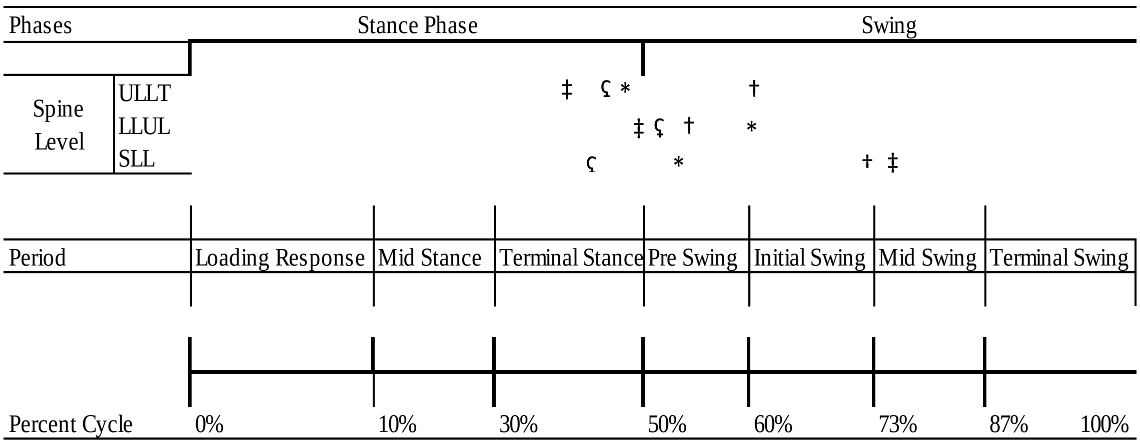


Figure 9. Where frontal plane peak angles occurred during gait. * Denotes walking condition; † obstacle crossing; ‡ stair ascent; ʢ stair descent

There was not a significant interaction between spine level and task in the frontal plane range of motion ($p = 0.132$). There was a significant main effect of spine level for the ipsilateral rotation range of motion (Table 2). Pairwise post hoc comparisons of the spine level marginal means revealed SLL ($14.48^0 \pm 3.25^0$) ROM was significantly larger than LLUL ($10.96^0 \pm 3.71^0$). It should be noted that SLL was trending larger than ULLT ($6.31^0 \pm 1.21^0$; p=0.019) which is outside considered significance with the Bonferroni adjustment.

Table 2. Ipsilateral bending outcome measures of maximum peak angle, maximum peak angle index and range of motion.

	SLL	LLUL	UPLT	p-value
Maximum Peak Angle (deg) Mean (SEM)				
Spine Level (Overall)				.499
Task (Overall)				.088
Walking	5.19 ± 2.01	5.28 ± 3.30	5.79 ± 1.10	
Obstacle Crossing	10.09 ± 1.94	8.20 ± 3.70	7.64 ± 1.42	
Sair Ascent	13.48 ± 2.47	6.22 ± 1.92	10.68 ± 1.66	
Stair Decent	9.32 ± 3.02	5.90 ± 4.58	5.08 ± 0.98	
Maximum Peak Angle Index(%) Mean (SEM)				
Spine Level (Overall)				
Task (Overall)				
Walking	57.74 ± 4.06	59.99 ± 3.32	47.97 ± 4.78	
Obstacle Crossing	71.95 ± 3.65	55.67 ± 5.66	60.09 ± 6.02	
Sair Ascent	73.56 ± 3.70	46.40 ± 6.28	39.02 ± 5.28	
Stair Decent	47.43 ± 3.41	51.99 ± 5.61	44.25 ± 7.18	
Range of Motion (deg) Mean (SEM)				
Spine Level (Overall)				.023
Task (Overall)				.077
Walking	9.94 ± 3.73	8.94 ± 4.57	2.59 ± 0.31	
Obstacle Crossing	16.16 ± 2.68	12.88 ± 4.59	6.04 ± 2.04	
Sair Ascent	15.79 ± 0.85	12.53 ± 1.32	13.79 ± 2.09	
Stair Decent	16.04 ± 5.73	9.50 ± 4.35	2.83 ± 0.39	

p: main effect Greenhouse-Geisser adjusted significance level.

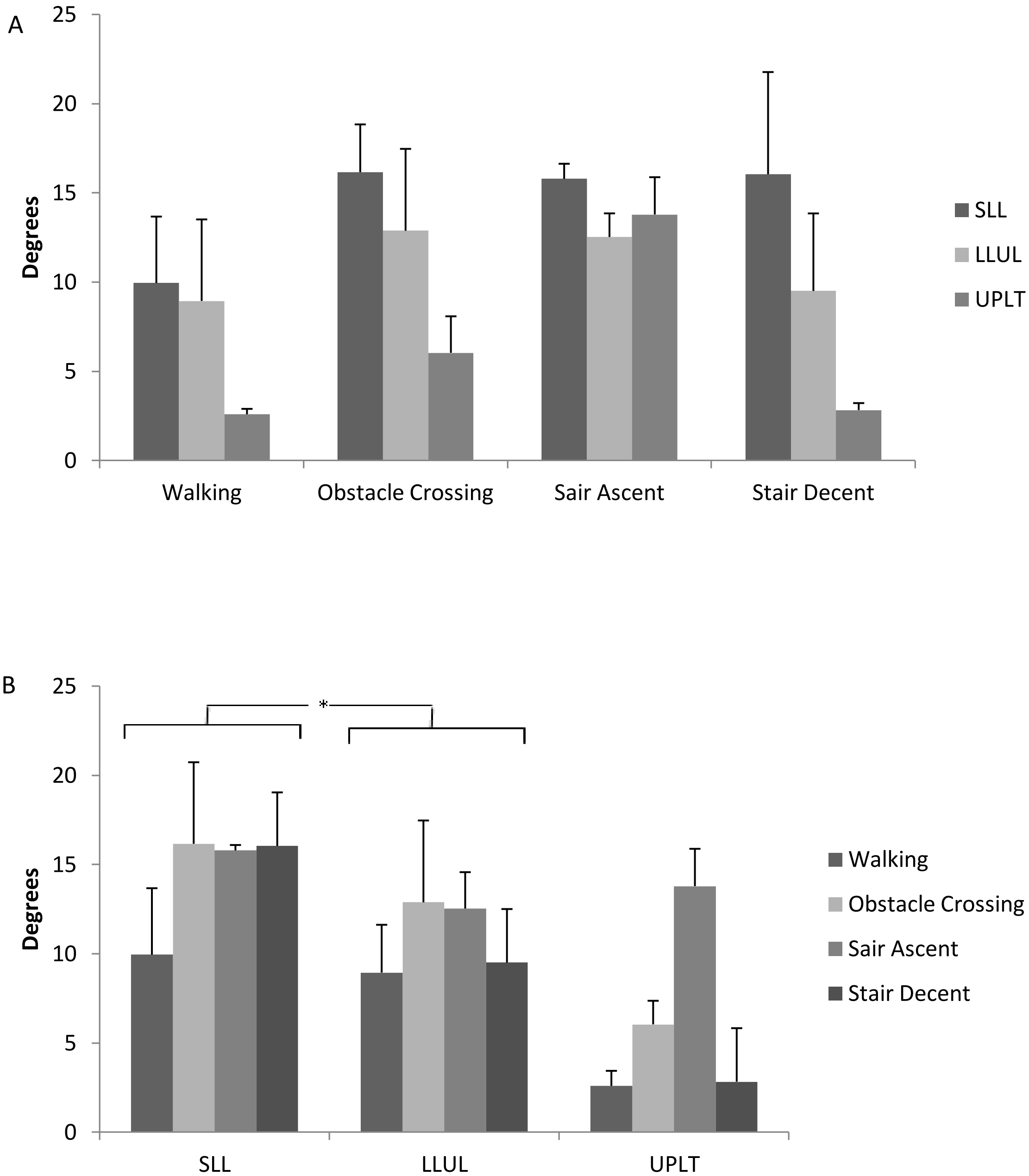


Figure 10. Range of motion in the frontal plane of motion for various activities of daily living. A) is spine level ROM for each task of daily living. B) is the task specific ROM for all spine levels.

Transverse Plane Motion

There was a significant interaction between spine level and task for the maximum peak angle in the transverse plane of motion ($p = 0.024$). Post hoc pairwise comparisons on the spine level marginal means did not find any individual group differences due to the Bonferroni adjustment. It should be noted that during the W task SLL was trend larger than ULLT (p=0.008) and during OC task SLL was trending larger than ULLT (p=0.01) [Figure 11].

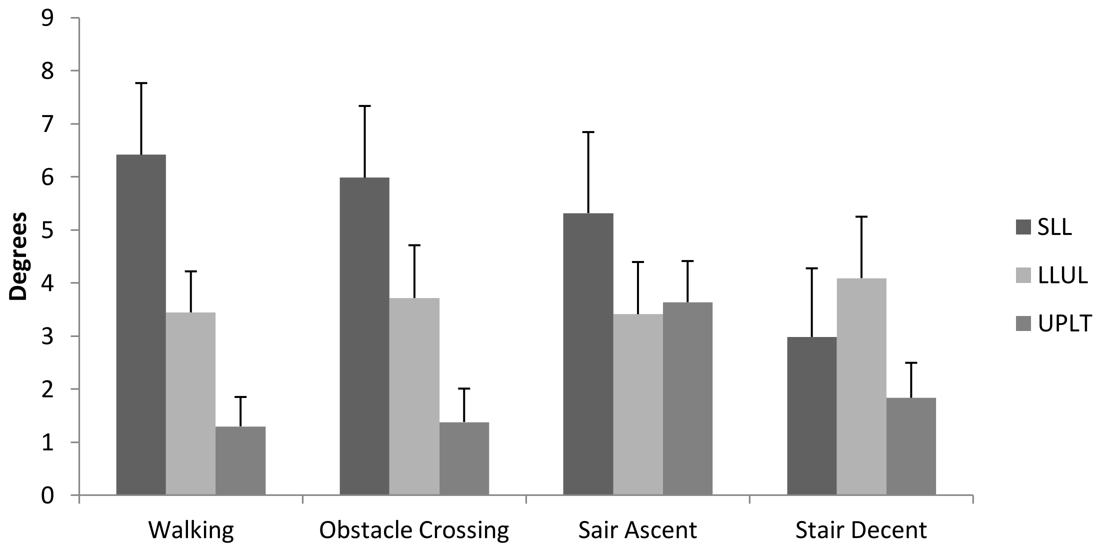


Figure 11. Peak contralateral axial rotation angle for various activities of daily living.

The peak contralateral axial rotation angle during walking for the SLL, LLUL and ULLT joints occurred during terminal swing, just before HS of the contralateral limb (~48% GC). Obstacle crossing and SA produced a peak contralateral rotation angle for the SLL, LLUL and ULLT joints between mid swing and initial HS of the contralateral limb (~35% GC). Stair descent yielded a peak contralateral rotation for the SLL, LLUL and ULLT joints between mid-swing and initial HS of the contralateral limb contacted the stair (~40% GC) [Figure 12].

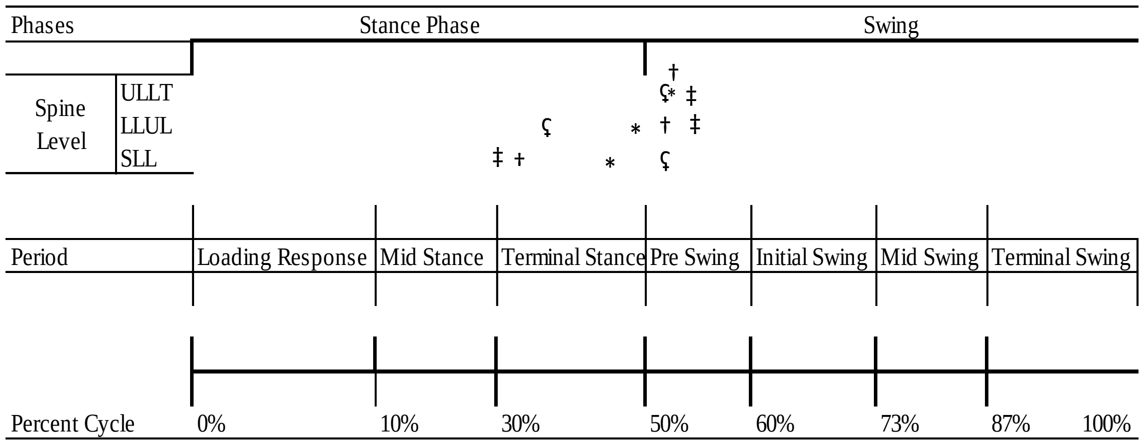


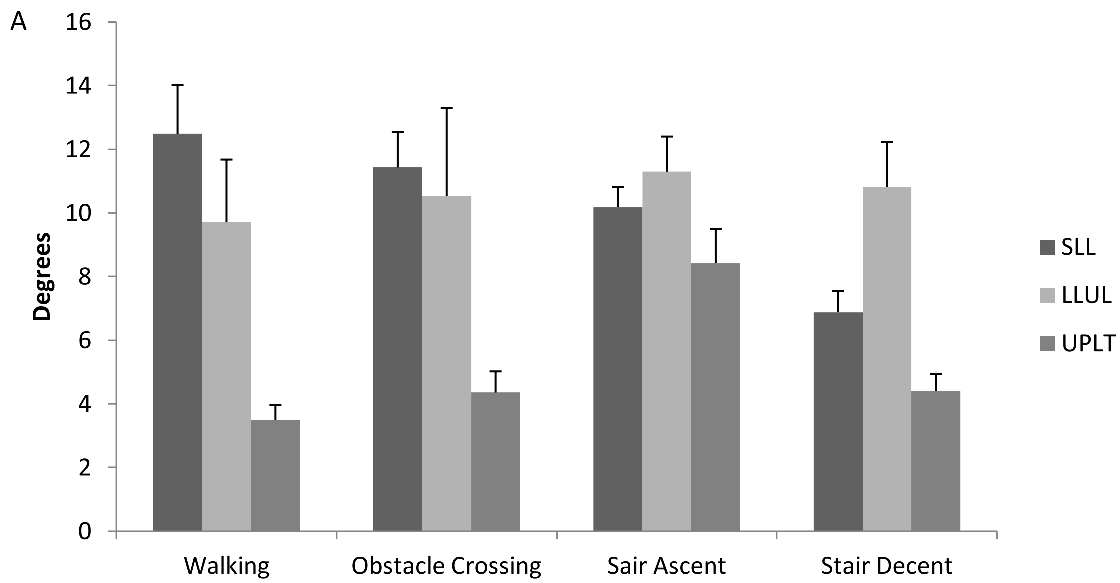
Figure 12. Where peak contralateral bending angles occurred during gait. * Denotes walking condition; † obstacle crossing; ‡ stair ascent; ʕ stair descent

There was a significant interaction between spine level and task for transverse plane range of motion in the transverse plane of motion (p = 0.004). Post hoc pairwise comparisons revealed for SLL spine level, the OC ($11.43^0 \pm 1.12^0$) task had significantly larger ROM than SD ($6.88^0 \pm 0.66^0$). Additionally, during W the SLL ($12.48^0 \pm 1.54^0$) joint produced significantly more ROM compared to ULLT ($3.49^0 \pm 0.48^0$). In addition, the OC task produced a significantly larger SLL ($11.43^0 \pm 1.12^0$) ROM compared to ULLT ($4.35^0 \pm 0.48^0$). The SD task yielded a significantly larger ROM in LLUL ($10.81^0 \pm 1.42^0$) than ULLT ($4.41^0 \pm 0.52^0$) [Figure 13b].

Table 3. Contralateral axial rotation outcome measures of maximum peak angle, maximum peak angle index and range of motion.

	SLL	LLUL	UPLT	p-value
Maximum Peak Angle (deg) Mean (SEM)				
Spine Level (Overall)				N/A
Task (Overall)				N/A
Walking	6.42 ± 1.35	3.45 ± 0.77	1.30 ± 0.55	
Obstacle Crossing	5.99 ± 1.35	3.72 ± 0.99	1.38 ± 0.63	
Sair Ascent	5.32 ± 1.53	3.41 ± 0.98	3.64 ± 0.78	
Stair Decent	2.98 ± 1.29	4.09 ± 1.16	1.84 ± 0.66	
Maximum Peak Angle Index(%) Mean (SEM)				
Spine Level (Overall)				
Task (Overall)				
Walking	44.87 ± 4.26	48.16 ± 3.47	52.37 ± 4.09	
Obstacle Crossing	30.38 ± 4.05	52.50 ± 5.99	51.96 ± 4.51	
Sair Ascent	27.81 ± 4.22	56.49 ± 5.03	54.61 ± 5.21	
Stair Decent	52.44 ± 5.63	38.13 ± 5.11	52.10 ± 5.60	
Range of Motion (deg) Mean (SEM)				
Spine Level (Overall)				N/A
Task (Overall)				N/A
Walking	12.48 ± 1.54	9.70 ± 1.97	3.49 ± 0.48	
Obstacle Crossing	11.43 ± 1.12	10.53 ± 2.78	4.35 ± 0.67	
Sair Ascent	10.18 ± 0.64	11.29 ± 1.10	8.42 ± 1.07	
Stair Decent	6.88 ± 0.66	10.81 ± 1.42	4.41 ± 0.52	

p: main effect Greenhouse-Geisser adjusted significance level.



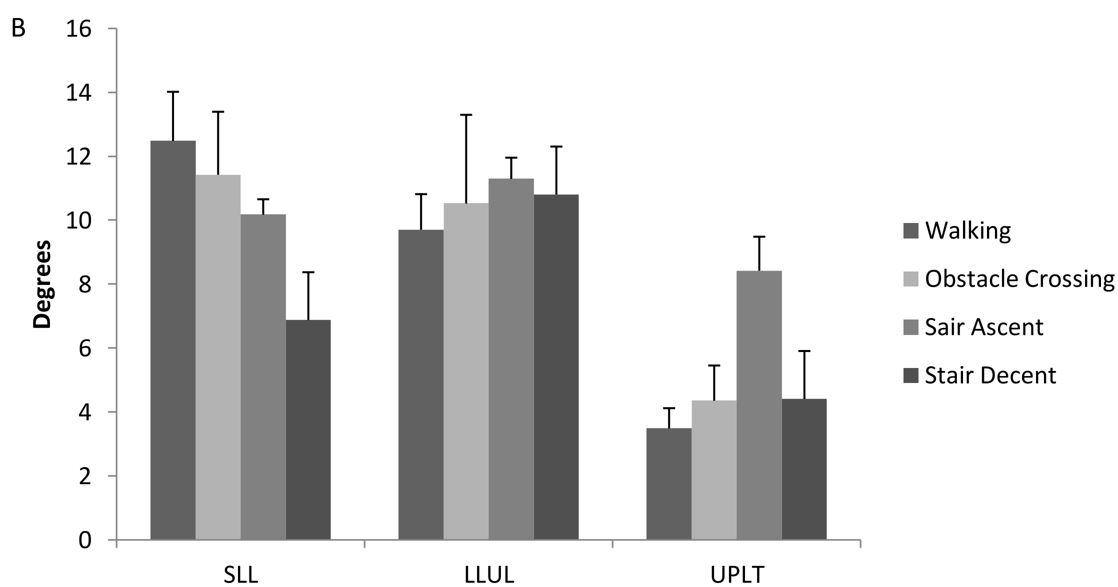


Figure 13. Range of motion in the transverse plane of motion for various activities of daily living. A) is spine level ROM for each task of daily living. B) is the task specific ROM for all spine levels.

Discussion

The purpose of this study was to examine the effect of different activities of daily living (level walking, obstacle crossing, stair ascent and descent), on kinematics of individual spinal segments in young adults. It was hypothesized that different activities of daily living will exhibit different kinematics at different segmented spinal segments.

Significant interactions for the contralateral axial rotation peak angles and ROM were detected. Post-hoc pairwise comparisons of the marginal means (spine level and task) values and individual factor values were not significant. Main effects for spine were found for peak flexion angle, sagittal ROM and ipsilateral bending. Furthermore, main effects on task were observed in peak flexion angle and sagittal ROM. Post-hoc pairwise comparisons on spine level marginal means found significant differences in the sagittal ROM (SLL > ULLT and LLUL > ULLT), frontal plane ROM (SLL > LLUL) and transverse plane ROM (SLL > ULLT and LLUL > ULLT). Finally, specific significant differences were recorded in the ULLT frontal plane of motion spine level in which it was observed that the SA task had a greater ROM than the W task. Additionally, the SA task also had more ROM than the SD at the ULLT frontal spine level. In the transverse plane of motion, the SLL spine level presented with significantly larger OC ROM than the SD task. Overall, the results supported the hypothesis of this study.

The SLL joint had the largest ROM in the sagittal plane of motion. This observation may be explained by the anatomic location of the fifth lumbar vertebra and the unique spatial orientation of the facet articulation with the sacrum (White & Panjabi, 1990). In addition to the orientation of the facet joints, the SLL joint has the largest intervertebral disc of the entire spine. White and Panjabi (1990) have shown the size of the intervertebral discs strongly correlates with the ROM available at that joint. Furthermore, the large ROM in the SLL joint may be explained by the body positioning needed to develop momentum to complete the task. This is done by positioning of the whole body COM closer to the anterior boundary of the base of support (BOS). This body orientation will provide a larger forward moment created by the COM and will assist in the forward momentum of the body (D. A. Winter, 2009). This may explain why the SA task has larger peak flexion angles than SD. The SA task requires momentum to be generated to facilitate ascending, while the SD task is assisted by gravity. During the SD task, the spine will be keeping the COM will within the BOS to maintain balance.

Results from the current study indicate that the SLL ROM was larger than the ULLT in the transverse plane of motion and trending larger than LLUL (p=0.01). This indicates more motion in the spine produced in the inferior segments and suggests the reason most back injuries are located in the inferior region are due to an overuse degenerative properties associated with increased ranges of motion, larger moment production is associated with an increased motion (NINDS, 2011; White & Panjabi, 1990). A larger moment would induce more stresses on the tissues in the low back area and an increased stress may indicate a greater chance for injury. However, results of this study seem to contradict basic anatomy. Orientation of the facet joints suggest that the majority of rotation should occur in the thoracic spine. However, in the current study, the SLL joint had significantly more rotation ROM than ULLT. This might suggest during gait-related tasks, the movement of the lower extremity may influence the motion of the spine and thus counteract the intended motion of the spine. Furthermore, the costal bones (rib cage) and sternum might also limit rotation of the thoracic spine in order to maintain the integrity of the lungs during walking.

Most of the peak angles occurred during, or very close to, the weight acceptance phase of the contralateral foot. The maximum segmental spinal angles are possibly present during this phase of task because this is when the most forward momentum from the trunk and spine needs to be produced. This production of momentum from the spine may be needed to assist in moving the body forward to overcome the breaking force, which is applied to the COM that occurs with heel strike (D. A. Winter, 2009). The momentum generated by the trunk will assist the trailing limb plantar flexors by generating the required force to keep the body moving (Perry, 1992).

This was the first study to investigate spinal motion in this detail during ambulatory tasks. Thus, many follow up comparisons were run between segmented spine level and task to examine the hypothesis. Controlling against a false positive (Type I error) required adjustment to the significant level. Although several post hoc pairwise comparisons were found to have $p<0.05$; with the large inter-individual variability, it was not surprising that several of the factors were observed as not significant. Conversely, the effect size (ES) of many of the factors, represented by partial eta squared η^2) were medium (0.06) to large (0.14). The ES for all factors ranged between 0.044 (ipsilateral bending ROM spine level) to 0.522 (contralateral axial rotation ROM spine level). All ES were above 0.185 except, ipsilateral bending ROM spine level and contralateral axial rotation peak angle task. This suggests that although the observed changes in the dependent variables (peak angle and ROM) are small in magnitude, the cumulative effect of these slightly different segmented spinal motions between ambulatory activities of daily living may become important in understanding back pain and injury.

Findings from this study could be used to develop new treatment protocols for individuals with back pain. It has been reported that back pain is most commonly located in the lower areas of the spine (NINDS, 2011). The results showed most of the motion in the spine to occur in the inferior segmented spine joints, suggesting more movement in the low back will lead to more injuries (White & Panjabi, 1990). To limit the lower spine in its movement, individuals who have back pain may be trained to utilize different (more superior) areas of the spine to produce the required movement in order to complete a task. Additionally, this sort of training could also be used as a preventative method to reduce the occurrence of back pain in the population.

One limitation of the present study could be due to the large inter-subject variability of the multi-segmented spine biomechanical outcome parameters. The sample size may have not been sufficient to fully observe differences in segmented spine motion during ambulatory activities of daily living. Nevertheless, the data was able to direct multiple motion patterns for different activities of daily living. Further studies might consider the incorporation of erector spinae activity during investigations of detailed spine motion.

In conclusion, the results from his study supported the hypothesis that different activities of daily living exhibited different kinematics (peak angle and range of motion) at different spinal segments. Changes in the sagittal plane might suggest momentum generation in a way to help complete the task or maintain balance depending on the task. Greater ranges of motion in the more distal segments (SLL) than the proximal (LLUL and ULUT) could provide evidence as to why most injuries occur in the low back as opposed to the middle or upper back. Overall, this study was able to show how a segmental spine marker set could be an effective tool in determining different motion patterns from various spinal segments during multiple activity of daily living. It should be noted that some further validation may be appropriate with a larger and more diverse healthy population before applying this procedure to unhealthy individuals.

Bridge

Chapter III shows that different ambulatory activities of daily living will induce a change in segmental spinal mechanics. Changes in spine position as a function for different tasks of daily living suggest the spine moves differently in response to the different muscle recruitment patterns during those activities; or moves differently depending on the ROM required at the distal joints during those activities. The data from this study could be foundational work later utilized to develop back pain treatment protocols which could suggest postures which may alleviate pain during movement, thereby keeping people active following the appearance of back pain. Chapter IV investigates the relationship between age (individuals between 20-59 years) and spine kinematics during gait and stair descent.

CHAPTER IV

AGE EFFCTS ON SPINE MOTION DURING AMBULATORY ACTIVITIES

This chapter was developed by Li-Shan Chou, Ph.D. & Scott P. Breloff. Dr. Chou contributed substantially to this work participating in the development of methodologies and providing invaluable critiques and substantial editing advice. Scott P. Breloff was the primary contributor to the development of the protocol, data collection, data analysis and did the writing.

Introduction

As baby boomers reach their 60s, the US population is aging at a rapid rate. Similarly, individuals 55 and older are the fastest-growing segment of the population (Gfroerer, Penne, Pemberton, & Folsom, 2003). Most back pain occurs in individuals between thirty and fifty years (NINDS, 2011). A recent systemic review of the global prevalence of back pain found the highest occurrence of back pain was in individuals between forty and eighty years old (Hoy et al., 2012). With the increasing age of the population and the potential for back pain to affect individuals to such a mature age, back pain can place a financial burden on the healthcare system (Rogers, 2001). The frequency of back pain and the lack of treatment methods have motivated many investigators to focus on back pain related research (Bigos et al., 1994). It is the hope that results from these studies could enhance the understanding of possible biomechanical mechanisms which lead to back pain, thereby aiding in the development of better treatment methods and new preventative procedures.

Age-related changes have been reported in balance control during locomotion such as the center of mass motion during obstacle crossing and medio-lateral stability(Fiatarone & Evans, 1993; JudgeRoy et al., 1996). Various spatiotemporal age related gait parameter change such as, decreases in stride lengths, increases in step widths, and deceases in medio-lateral stability during walking have been reported (Fiatarone & Evans, 1993; Hahn & Chou, 2004; Hollman et al., 2007; JudgeRoy et al., 1996; Maki, 1997; Menz et al., 2003; Schrager et al., 2008). These changes observed in the lower extremity could be related to alterations in spine/trunk movement control due to the aging process. Older adults were reported to have greater angular sway and trunk velocities in the sagittal plane when compared to young and middle-aged individuals during stance and level walking (Gill et al., 2001). Additionally, older and younger individuals have different trunk acceleration characteristics during normal gait (J. J. Kavanagh et al., 2005; J. Kavanagh et al., 2004).

In addition to changes in level walking, ageing has shown to drastically increase the injuries which occur during staircase negotiation (Hemenway, Solnick, Koeck, & Kytir, 1994). Age-related changes were reported in the trunk range of motion during stair negotiation (McGibbon & Krebs, 2001). It has been found that pelvic rotations in sagittal, frontal and transverse planes of motion were systematically reduced with age (Van Emmerik, McDermott, Haddad, & Van Wegen, 2005). Furthermore, aging also produces reductions in the passive range of motion of the trunk, as older individuals were found to have a smaller range of motion in the trunk (flexion/extension, bilateral side bending, and bilateral axial rotation) in a group of individuals from 20 years to 60+ years (Van Herp, Rowe, Salter, & Paul, 2000). Furthermore, it has been reported that significant reduction in spinal range of motion exists with advancing age (Crosbie, Vachalathiti, & Smith, 1997a).

The previous studies used large segment to define the back and were mostly interested in the relative movement between the pelvis and spine. This limits the information regarding individual spinal segment motion during activities of daily living in individuals of different decades of life. Thus, the implementation of an in-vivo quantification approach to detect detailed spine motion at multiple segment levels would further the understanding of spine motion.

Quantifying the motion between adjacent spinal segments in individuals during different decades of life will allow for a more thorough understanding of age-related changes in the spinal motion. A more specific detection on changes in the spine motion would better allow a better understanding of the mechanisms which are present as a result of aging. A better understanding may lead to better treatment protocols and new preventative procedures to help individuals who suffer from back pain. Thus, the purpose of this study was to investigate whether individuals in consecutive age groups (20-29, 30-39, 40-49, and 50-59 years) will display altered spinal kinematics during level walking and stair descent. It was hypothesized that individuals in different age groups will exhibit particular kinematics changes at the specific spinal segments during level walking and stair descent.

Methods

Subjects

Thirty-five healthy adults subjects, demarcated by four age groups: 20-29 years (6 men/4 women; mean age: 24.10±2.64 years, mean height: 181.58±33.53 cm, mean mass: 65.35±13.33 kg), 30-39 years (3 men/4 women; mean age: 33.43±3.14 years, mean height: 168.68±8.99 cm, mean mass: 71.74±19.70 kg), 40-49 years (5 men/6 women; mean age: 45.09±2.43 years, mean height: 168.45±4.96 cm, mean mass: 65.69±8.01 kg), 50-59 years (2 men/4 women; mean age: 53.14±2.91 years, mean height: 164.47±9.77 cm, mean mass: 69.23±16.90 kg), were recruited from the university and surrounding community to participate in the study. Subjects did not have a history or clinical evidence of neurological, musculoskeletal or other medical conditions affecting gait performance, such as stroke, head trauma, neurological diseases (i.e. Parkinson's, diabetic neuropathy), visual impairment (which could not be corrected by lenses) and dementia. All subjects reviewed and signed an informed consent approved by the Institutional Review Board.

Experimental Protocol

Male subjects were asked to wear spandex shorts with no shirt. Female subjects wore a dance leotard with open back. Both male and female subjects performed two different tasks while with bare feet: level ground walking (W) and stair descent (SD). The task order was randomly selected for each subject, and the total protocol duration was not extensive enough to induce fatigue (Yoshino, Motoshige, Araki, & Matsuoka, 2004). The level walking task required subjects to walk along a 10 meter long walkway. When performing stair descending, subjects were instructed to initiate their SD trials from the back end of an elevated walkway; they then descended a 3-step staircase, and continued walking for several steps (Figure 1). Subjects were instructed to perform both tasks at their usual walking paces. Motion data from five trials of each task were collected from each subject for analysis.

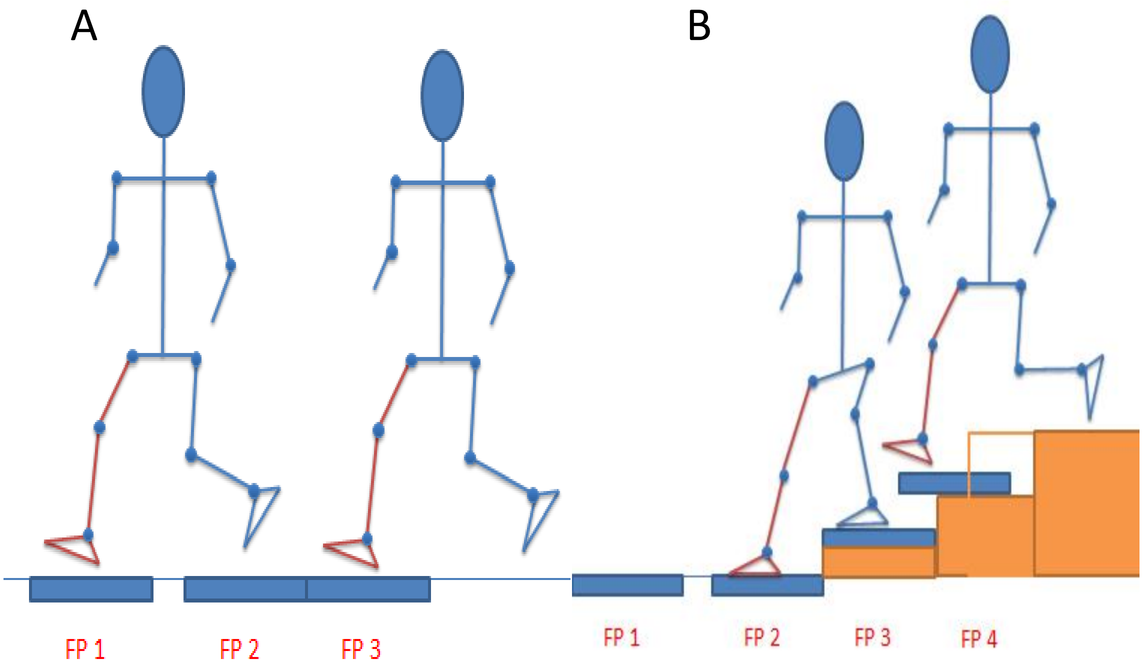


Figure 1. Definition of each task. (A) Level walking (W) – ipsilateral heel strikes, (B) Stair Descent (SD) - ipsilateral heel strikes.

Whole body motion analysis was performed with a ten-camera motion analysis system (Motion Analysis Corporation, Santa Rosa, CA). Sixty-two retro-reflective markers (diameter=14mm) were placed on the subject. In addition to a whole body marker set (Hahn and Chou, 2004), eight markers were placed directly on the palpated spinous processes of the following vertebrae; Cervical 7 (C7), Thoracic 3 (T3), Thoracic 6 (T6), Thoracic 9 (T9), Thoracic 12 (T12), Lumbar 3 (L3), Sacrum 1 (S1) & Sacrum 5 (S5). Two markers were placed on the left and right posterior superior iliac spine (PSIS), and the remaining markers were placed 50mm to the left and right of the spinous process markers, except for S1 and S5 (Figure 3). Three-dimensional marker position data were collected at 60 Hz and low-pass filtered using a 4^{th} order Butterworth filter with the cutoff frequency set at 5 Hz.

Three force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were placed in series and embedded level into the laboratory floor for the walking (W) task. The first two force plates were immediately adjacent to one another, and the third plate was separated by a distance of 15cm. This setup was to accommodate subjects walking with different step lengths. Gait events such as heel strike (HS) and toe off (TO) were detected using the vertical ground reaction force (GRFv). Heel strike was determined to occur when the GRFv was greater than 10% of the maximum GRFv, and toe off was determined to occur when the GRFv was less than 10% of the maximum GRFv (Ghoussayni et al., 2004; Hreljac & Marshall, 2000; Mickelborough et al., 2000).

For the stair descending (SD) task, a three step staircase was used. Each step had a rise of 17.8 cm, a run of 30.5 cm and a width of 80 cm, forming a stair angle (rise/run) of 30^0 (H. J. Lee & Chou, 2006) A total of four force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were used to obtain ground reaction force data during SA and SD trials. Two force plates were embedded level into the laboratory floor and two made up the steps (see Figure. 2, Chapter III).

Data Analysis

Marker position data were analyzed for one task cycle during each trial, and five trials were examined for each task from each subject. A gait cycle during level ground walking was defined as the time interval between two consecutive ipsilateral heel strikes (FP 3 to FP 1; Figure 1a) and stair descent was examined consecutive ipsilateral heel strikes following first step down to the ground level (FP 4 to FP 2; Figure 1b)

A MATLAB® (Mathworks, Natick, MA) program was developed to calculate six adjacent segmental spinal angles. In this study, only the three most inferior adjacent spine angles were presented due to the fact that most back pain presents in the low back (NINDS, 2011). The angles in the current study are labeled as: sacrum to lower lumbar [SLL], lower lumbar to upper lumbar [LLUL] and upper lumbar to lower thorax [ULLT] (see Figure. 1, Chapter III).

Peak angles, timing of the peak angle and range of motion (ROM) have been previously used to describe spinal motion during ambulatory activities of daily living (Jorrakate et al., 2011; Mayer et al., 1984; McLean et al., 2005; Neblett et al., 2010; Perry, 1992; Samuel et al., 2011; Varin et al., 2011; D. A. Winter, 2009). Additionally, ROM has been used to detect aging effects in the spine during gait (Fitzgerald, Wynveen, Rheault, & Rothschild, 1983; Mayer et al., 1984; Sullivan & Dicknison, 1994; Yamamoto et al., 1989). In the current study, peak spine angle and ROM were examined at different spine levels and during four distinct (level walking, obstacle crossing, stair ascent and stair descent) tasks of daily living of individuals in different age groups. Peak angles were recorded for the three different planes of motion and the peak flexion angle was presented for the sagittal plane. Spine flexion is forward/flexion bending of the spine. In the frontal plane, peak ipsilateral bending is discussed, which indicates the spine is bending toward the initial contact leg, or the ipsilateral side. In the transverse plane, the motion discussed is peak contralateral axial rotation. This motion requires the spine to bend away from the striking leg or toward the contralateral leg. In the current study, data analysis was over one gait cycle, as described in Figure 1. The foot which first contacted the force plate was then the reference for ipsilateral and contralateral. When appropriate, motion in the frontal and transverse plane were multiplied by negative one to normalize the left and right sides. The timing of the peak angle (index) is the occurrence of the peak angle during the task. In the current study, this biomechanical parameter is used to describe the gait event that is occurring during the peak angle in each plane of motion. Range of motion is the total excursion of the segmental joint angle during each task.

Peak angles and ranges of motion were analyzed using a three-way, mixed-effects analysis of variance. Although there were different numbers of subjects in each group (due to recruitment and data quality), any misrepresentation of the data was alleviated by using the Type III Sum of Squares in SPSS. The Type III Sum of Squares adjusts the harmonic mean of the unbalanced design to assure acute interpretation of the data (Maxwell & Delaney, 1990). Age was an independent between-subject factor with four levels: (a) 20-29 years, (b) 30-39 years, (c) 40-49 years, and (d) 50-59 years. The second factor was the spinal joint level, a within-subject factor, with three levels: sacrum to lower lumbar (SLL), lower lumbar to upper lumbar (LLUL) and upper lumbar to lower thorax (ULLT). Task was also a within-subject factor with two levels: level waking and stair descent. In all cases adjusted p-values (Greenhouse-Geisser) were used to evaluate within subject effects because the assumption of sphericity was evaluated with the

Mauchly Sphericity Test and found to be non-tenable, $p < 0.05$. If the three-way interaction was not significant, then three two-way ANOVA's were run to investigate differences between factors. If a significant interaction between factors was detected, pair-wise and main effect comparisons were applied using the Bonferroni method to identify the differences. When the interaction was not significant then only the main effects of each factor would be discussed. The level of significance for these statistical tests was set at 0.05. Additionally, partial eta squared (η2) were calculated as effect sizes (ES) for all variables to assist in the explanation of any trends. An ES was considered small (0.01), medium (0.06), and large (0.14) respectively (Cohen, 1988). This is interpretation of the eta squared statistic is appropriate because it is used to as an index of the strength of association between an independent variable (spine level and task) and a dependent variable (ROM and peak angles) that excludes variance produced by other factors (Pierce et al., 2004). The statistical software PASW (version 18, IBM., New York, NY) was used for all statistical analyses.

Results

All nine 3-way (i.e., age x spine level x task) ANOVA results were found to have non-significant interactions (Table 1). Therefore, three 2-way interactions (spine level * task, spine level * age and task * age) were interpreted for each biomechanical variables of interest in each plane of motion (Table 2, 4, and 6).

Table 1. *p*-values for 3-way ANOVA of each dependant variable.

	Peak Angle	ROM
Sagittal Plane	.804	.809
Frontal Plane	.055	.316
Transverse Plane	.179	.722

Sagittal Plane Motion

As there was no 3 way interaction, 2 way interactions (age x spinal level, age x task, spinal level x task) were examined. There was not a significant two-way interaction between the spine level and age factors. Additionally, there was not a significant two-way interaction between the task and age factors. A significant two way interaction was found for spine level and task (Table 2). Posthoc pairwise individual cell mean comparisons were unable to detect significant differences between spine level and task with Bonferroni adjustment (Figure 2). Finally, there was not a significant main effect of age on peak sagittal plane flexion angle.

The timing of the peak flexion angle was used to determine what gait events are occurring when segmental spine peak flexion is produced (Table 3). In all age groups (20's, 30's, 40's and 50's) the peak flexion angle at the SLL joint during walking occurred approximately just before contralateral HS into the weight acceptance phase on the contralateral limb (~45%-51% gait cycle [GC]). The LLUL joint produced peak angles for all age cohorts during weigh acceptance and into mid-stance of the contralateral foot (~55%-65% GC). In the ULLT joint during walking, the twenty, thirty and forty year old clusters found the maximum flexion angle to occur at the end of the heel rocker into the ankle rocker of the ipsilateral leg (~57% GC).

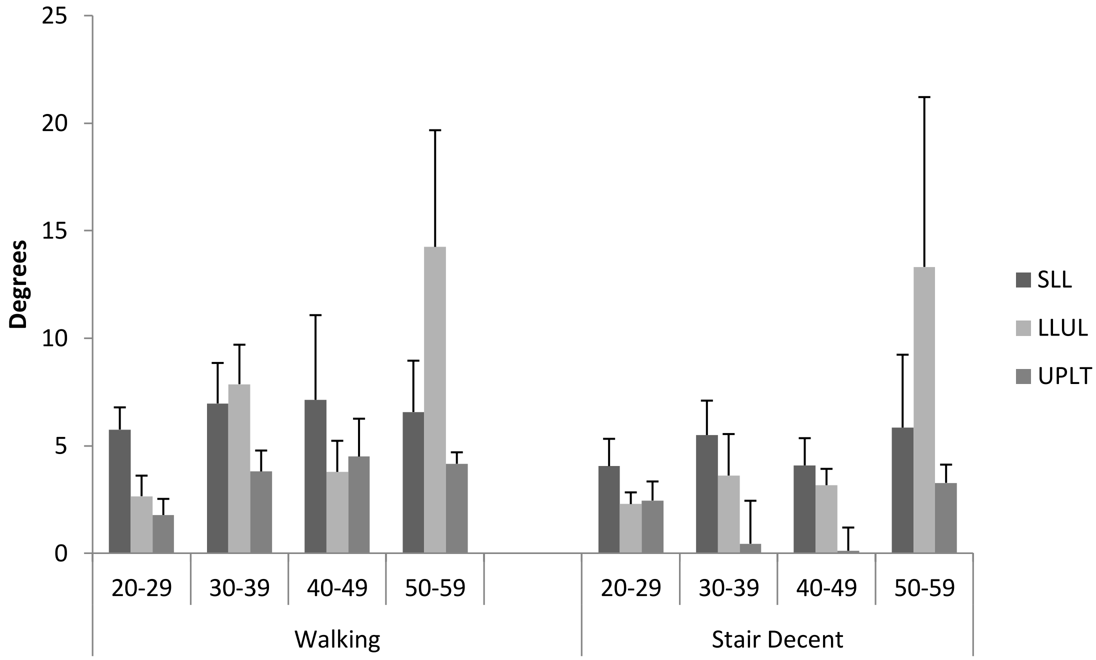


Figure 2. Flexion maximum peak values, different age groups and both tasks of daily living are on the horizontal axis.

Table 2. *p*-values for sagittal plane 2-way interactions

	p-value
Maximum Peak Angle	
Spine * Task	.040
Spine * Age	.677
Task * Age	.455
Range of Motion	
Spine * Task	.094
Spine * Age	.649
Task * Age	.848

*p:*significance value for the two-way ANOVAs in the sagittal plane

The fifty year old cohort had the maximum flexion angle during mid-stance (foot-flat) of the contralateral leg.

For the SD task and in the twenty and thirty year old cohorts, the maximum flexion angle presented during mid-swing to just before HS of the contralateral limb

(~40% GC). The older cohort of subjects (40 and 50 years old) had a maximum peak flexion in the SLL joint during mid-stance of the ipsilateral foot (~23-28% GC). In the LLUL joint, all age groups had a maximum flexion angle just before the contralateral foot contacted the step to mid-stance of the contralateral foot (~37% GC). In the ULLT joint, the maximum flexion angle for the 20’s 40’s and 50 year old groups occurred during weight acceptance until just before foot-flat on the contralateral leg (~56% GC). The thirty year old group produced a maximum flexion angle during mid-stance of the ipsilateral leg (~38% GC) [Figure 3].

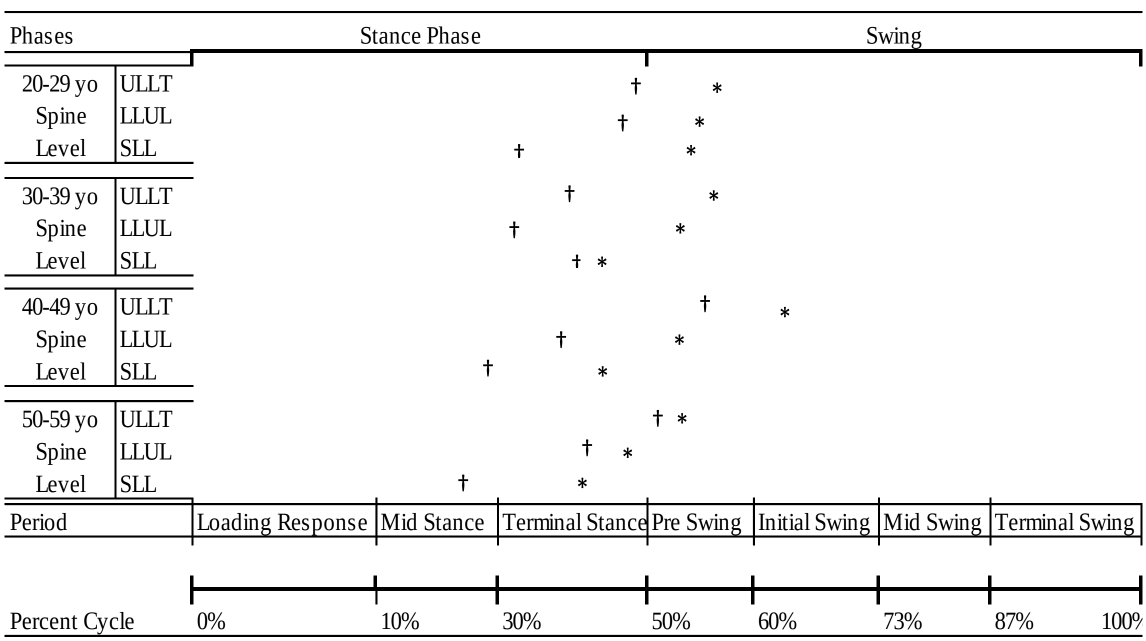


Figure 3. Timing of peak flexion angles for all age groups and both tasks. * Denotes walking condition; † stair descent

As there were no 3-way significant interactions, two way interactions (all combinations) were examined and all were found to be non-significant (Table 2). There was a significant main effect of spine level (p<0.001) for flexion range of motion. Follow up pairwise comparisons of the spine level marginal means found SLL (M=11.78±10.95) to have significantly more ROM than ULLT (M=4.93±2.17). Additionally, it was found that LLUL (M=9.42±7.56) had significantly more ROM than ULLT (M=9.42±7.56) [Figure 4B]. There was a significant main effect of task on sagittal ROM (*p*=0.006). When compared to W (M=10.58±3.05) the SD task was significantly smaller (M=6.71±1.14). There was not a significant main effect of age on sagittal ROM.

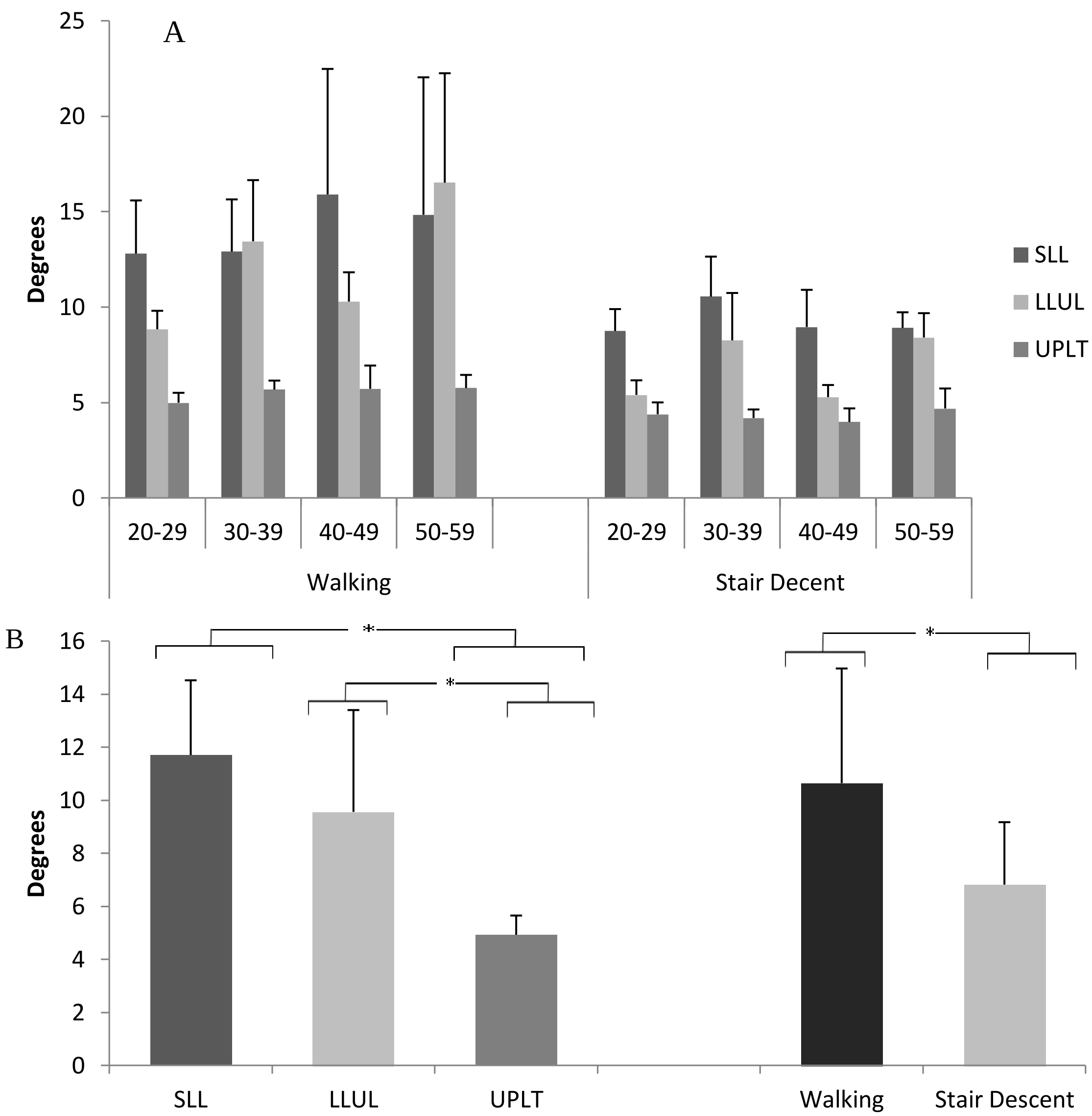


Figure 4. A) Flexion range of motion values; different age groups and both tasks of daily living are on the horizontal axis. B) Main effects of spine level and task.

Frontal Plane Motion

As there were no significant 3 way interactions, 2 way interactions (age x spinal level, age x task, spinal level x task) were examined and found to be non-significant (Table 1; Table 4; Figure 5). Additionally, main effects for spine level, task and age were not significant (p>0.05).

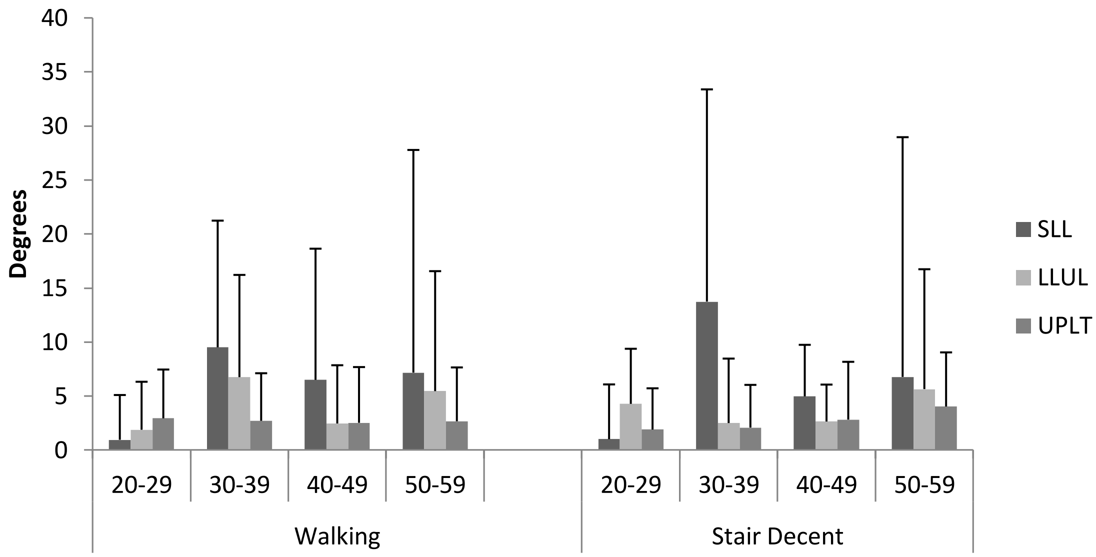


Figure 5. Ipsilateral bending maximum peak values; different age groups are listed the horizontal axis. Spine level angles are labeled in solid black for walking task and gray for stair descent.

For the walking task, the all age groups (20, 30, 40, and 50 years old) had a maximum peak ipsilateral bending angle during terminal swing into the loading phase on the contralateral foot at the SLL joint (~54% GC). In the LLUL joint, all age cohorts presented with a maximum ipsilateral bending angle during terminal swing an into initial contact loading on the contralateral limb (~53% GC). The ULLT joint produced a peak ipsilateral bending angle for all age groups between initial HS of the contralateral foot until mid-stance of the contralateral foot.

Table 3. Flexion maximum peak angle and ranges of motion for multiple spine levels during two ambulatory tasks of daily living.

Spine Level	SLL	LLUL	UPLT
Maximum Peak Angle (deg) Mean (SEM)			
Task (Walking)	6.58 ± 2.54	6.72 ± 2.38	4.16 ± 1.25
Task (Stair Decent)	4.72 ± 1.90	5.35 ± 1.89	3.28 ± 1.16
Overall	5.68 ± 2.26	6.06 ± 3.62	2.61 ± 1.20
Walking			
Age (20-29)	5.75 ± 1.03	2.66 ± 0.96	1.78 ± 0.75
Age (30-39)	6.97 ± 1.88	7.86 ± 1.84	3.81 ± 0.97
Age (40-49)	7.14 ± 3.94	3.79 ± 1.45	4.50 ± 1.76
Age (50-59)	6.57 ± 2.39	14.25 ± 5.43	4.16 ± 0.54
Stair Decent			
Age (20-29)	4.05 ± 1.27	2.29 ± 0.55	2.46 ± 0.88
Age (30-39)	5.50 ± 1.60	3.62 ± 1.93	0.44 ± 2.00
Age (40-49)	4.09 ± 1.27	3.17 ± 0.76	0.12 ± 1.08
Age (50-59)	5.85 ± 3.39	13.32 ± 7.90	3.28 ± 0.85
Range of Motion (deg) Mean (SEM)			
Task (Walking)	14.20 ± 5.10	12.03 ± 3.27	5.52 ± 0.78
Task (Stair Decent)	9.20 ± 1.43	6.64 ± 1.29	4.31 ± 0.70
Overall	11.78 ± 3.87	9.42 ± 2.67	4.93 ± 0.77
Walking			
Age (20-29)	12.80 ± 2.79	8.84 ± 0.97	4.99 ± 0.52
Age (30-39)	12.91 ± 2.74	13.44 ± 3.21	5.68 ± 0.47
Age (40-49)	15.90 ± 6.58	10.29 ± 1.54	5.72 ± 1.22
Age (50-59)	14.83 ± 7.20	16.52 ± 5.73	5.76 ± 0.69
Stair Decent			
Age (20-29)	8.75 ± 1.15	5.40 ± 0.77	4.38 ± 0.62
Age (30-39)	10.55 ± 2.10	8.26 ± 2.48	4.19 ± 0.45
Age (40-49)	8.96 ± 1.95	5.28 ± 0.64	3.98 ± 0.71
Age (50-59)	8.91 ± 0.81	8.41 ± 1.27	4.69 ± 1.05

During the SD task in the SLL joint, the twenty, thirty, forty and fifty year old cohorts all had a maximum ipsilateral bending angle between the mid-swing and terminal swing of the contralateral leg. The twenty and forty year old cohorts at the LLUL joint had a maximum ipsilateral bending angle during terminal swing of the contralateral limb (~43% GC). The thirty and fifty year old clusters had a maximum ipsilateral bending angle during mid-swing of the contralateral leg (~35% GC). The ULLT joint produced a peak ipsilateral bending angle between mid-swing to foot-flat on the contralateral leg in all age groups (Figure 6).

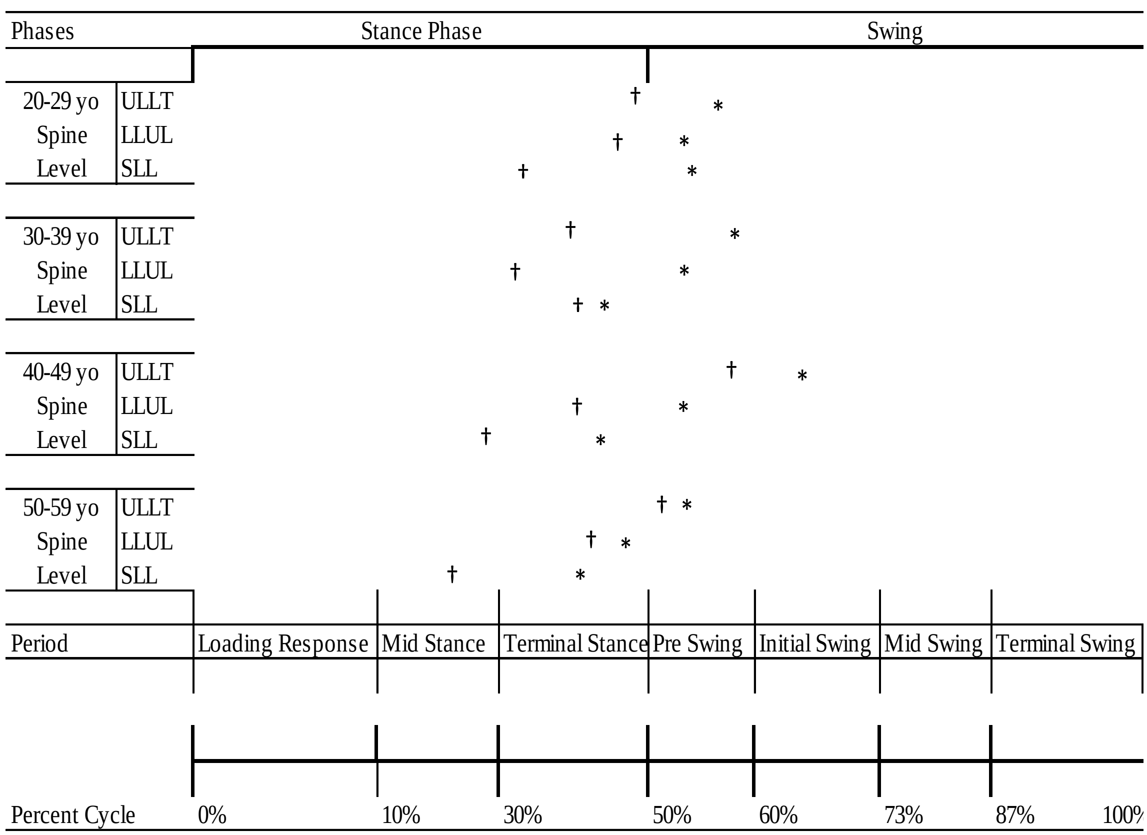


Figure 6. Timing of peak ipsilateral bending angles for all age groups and both tasks. * Denotes walking condition; † stair descent

Table 4. *p*-values for frontal plane 2-way interactions and pairwise comparisons

	p-value
Maximum Peak Angle	
Spine * Task	.301
Spine * Age	.351
Task * Age	.121
Range of Motion	
Spine * Task	.249
Spine * Age	.385
Task * Age	.287

*p:*significance value for the two-way ANOVAs in the frontal plane

There was not a significant three-way interaction between spine level, task and age for the ipsilateral bending range of motion (Table 1; p=0.316). Additionally, all three 2-way interactions for ipsilateral bending range of motion were not significant (Table 4). There was a significant main effect for spine level on ipsilateral bending (p=0.003). Post hoc pair wise comparisons of the spine level marginal means revealed SLL (M=10.81±2.47) was significantly larger than both LLUL (M=6.23±1.69) and ULLT (M=2.91±0.24) [Figure 7b]. There was not a significant main effect for task on the ipsilateral bending ROM (p>0.05). Furthermore, the main effect for age was not significant for ipsilateral bending (p>0.05).

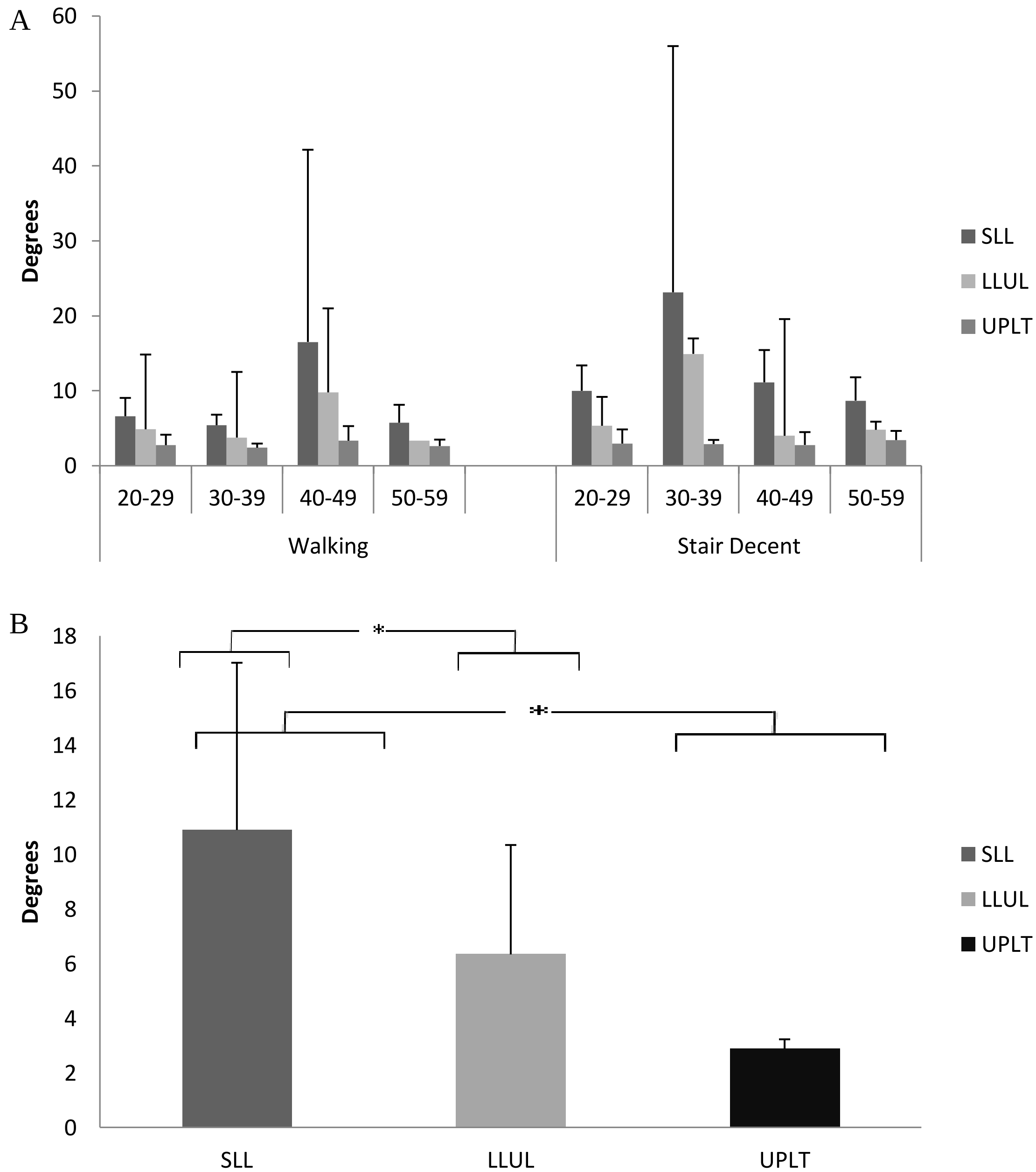


Figure 7. A) Ipsilateral bending range of motion; different age groups are listed the horizontal axis. Spine level angles are labeled in solid black for walking task and gray for stair descent. B) Main Effects of Spine Level for ipsilateral bending Range of Motion.

Table 5. Ipsilateral bending maximum peak angle and ranges of motion for multiple spine levels during two ambulatory tasks of daily living.

Spine Level	SLL	LLUL	UPLT
Maximum Peak Angle (deg) Mean (SEM)			
Task (Walking)	2.09 ± 2.39	3.85 ± 1.30	-1.38 ± 0.87
Task (Stair Decent)	2.80 ± 2.56	3.80 ± 1.11	-0.80 ± 0.87
Overall	2.43 ± 2.45	3.83 ± 1.20	-1.10 ± 0.86
Walking			
Age (20-29)	0.96 ± 1.46	1.88 ± 1.57	-2.94 ± 1.60
Age (30-39)	9.51 ± 4.79	6.75 ± 3.86	-2.71 ± 1.80
Age (40-49)	6.50 ± 4.29	2.46 ± 1.91	-2.53 ± 1.82
Age (50-59)	7.17 ± 7.29	5.45 ± 3.93	2.66 ± 1.76
Stair Decent			
Age (20-29)	1.01 ± 1.79	4.27 ± 1.80	-1.90 ± 1.35
Age (30-39)	13.74 ± 8.02	2.50 ± 2.44	-2.08 ± 1.61
Age (40-49)	4.97 ± 1.69	2.64 ± 1.21	-2.82 ± 1.89
Age (50-59)	6.77 ± 7.84	5.64 ± 3.92	4.03 ± 1.77
Range of Motion (deg) Mean (SEM)			
Task (Walking)	9.06 ± 2.40	5.72 ± 1.48	2.83 ± 0.23
Task (Stair Decent)	12.62 ± 2.54	6.76 ± 1.89	3.00 ± 0.25
Overall	10.81 ± 2.47	6.23 ± 1.69	2.91 ± 0.24
Walking			
Age (20-29)	6.60 ± 0.86	4.86 ± 1.36	2.74 ± 0.49
Age (30-39)	5.40 ± 0.57	3.75 ± 0.84	2.40 ± 0.23
Age (40-49)	16.51 ± 9.07	9.78 ± 5.50	3.34 ± 0.69
Age (50-59)	5.73 ± 0.85	3.33 ± 0.39	2.62 ± 0.30
Stair Decent			
Age (20-29)	10.00 ± 1.20	5.33 ± 0.71	2.95 ± 0.67
Age (30-39)	23.14 ± 13.41	14.92 ± 10.16	2.89 ± 0.22
Age (40-49)	11.13 ± 1.52	3.99 ± 0.26	2.77 ± 0.61
Age (50-59)	8.69 ± 1.10	4.79 ± 0.27	3.42 ± 0.43

Transverse Plane Motion

As the three-way interaction was not significant, two-way interactions (age x spinal level, age x task, spinal level x task) were examined. There was not a significant two way interaction for spine level and age or task and age; however a significant two way interaction for spine level and task was observed (Table 6; Figure 8). Post hoc pairwise comparisons spine level revealed that LLUL (M=4.14±0.82) was significantly larger than ULLT (M=1.59±0.59) during SD. The main effect for age on peak contralateral axial rotation angles was not significant (*p*=0.858).

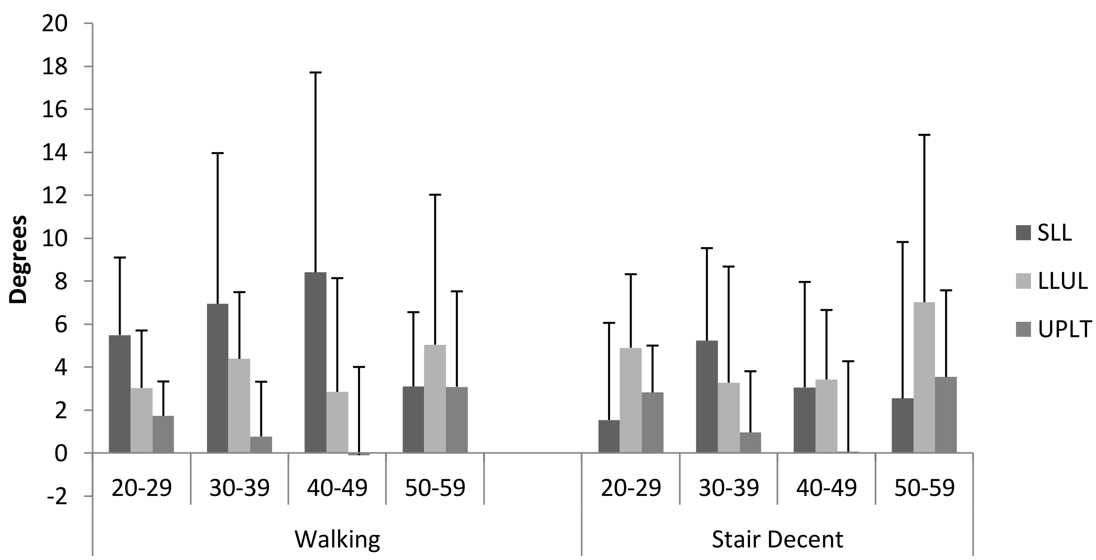


Figure 8. Rotation maximum peak angles; different age groups are listed the horizontal axis. Spine level angles are labeled in solid black for walking task and gray for stair descent.

During the walking task, the twenty and thirty year old cohorts had a maximum contralateral axial rotation angle during terminal swing of the contralateral limb at the SLL joint for all age groups (~45%GC). The twenty and thirty year old groups had a maximum contralateral axial rotation angle during initial contact of the contralateral leg in the LLUL and ULLT joints (~52% GC). The forty year old cohort's maximum contralateral axial rotation angle was found during terminal swing of the contralateral limb in the LLUL and ULLT joints (~47% GC). The fifty year old group's a maximum contralateral axial rotation angle in the LLUL and ULLT joints was during the heel rocker phase following contralateral HS (~56% GC).

During the SD task, the twenty, thirty, forty and fifty year old cohorts had a maximum contralateral axial rotation angle during the terminal swing to foot-flat of the contralateral leg at the SLL joint (~50% GC). In the LLUL joint during SD produced a maximum contralateral axial rotation angle during mid-swing to terminal swing of the contralateral limb in all age groups (~44% GC). The ULLT joint produced a peak ipsilateral bending angle between terminal swing to foot-flat on the contralateral limb (Figure 9).

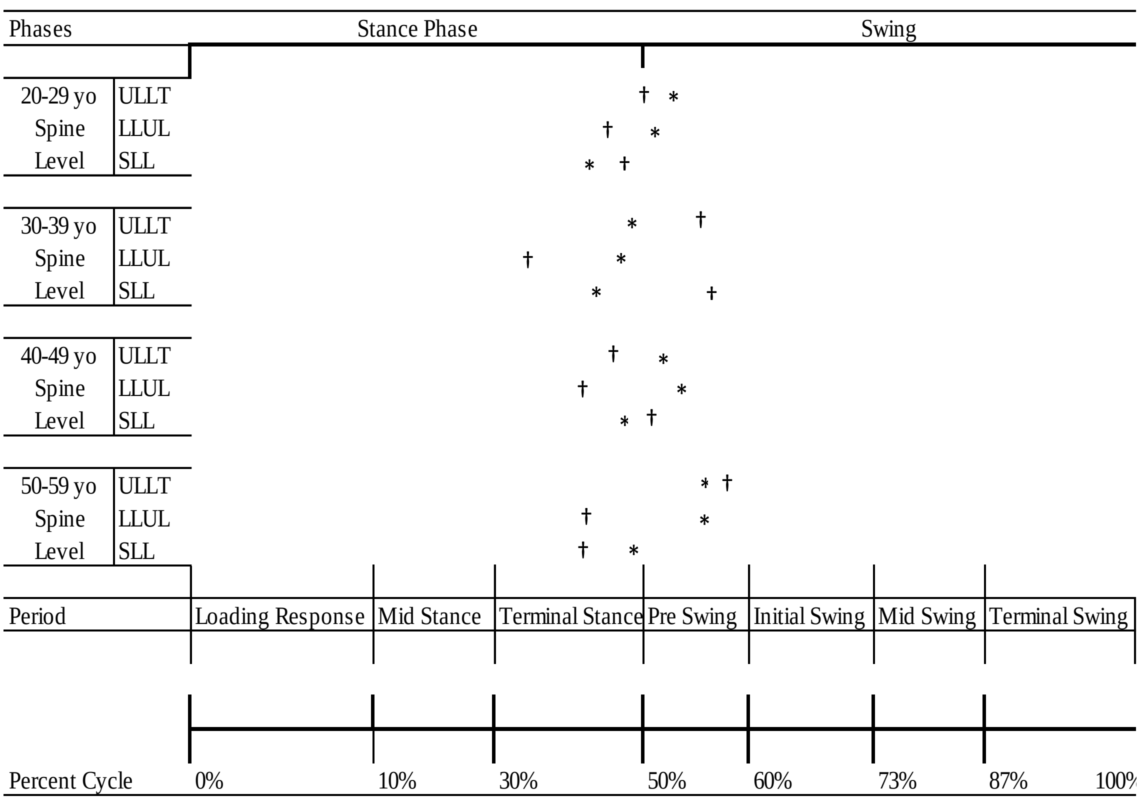


Figure 9: Timing of peak contralateral axial rotation angles for all age groups and both tasks. * Denotes walking condition; † stair descent

Table 6. *p*-values for transverse plane 2-way interactions and pairwise comparisons

		p-value
Maximum Peak Angle		
	Spine * Task	.001
	Spine * Age	.365
	Task * Age	.412
Range of Motion		
	Spine * Task	.001
	Spine * Age	.384
	Task * Age	.285

*p:*significance value for the two-way ANOVAs in the transverse plane

As there were no significant 3 way interactions, 2 way interactions (age x spinal level, age x task, spinal level x task) were examined for contralateral rotation ROM. The two way interaction between spine level and age was not significant. There was a non-significant two way interaction between task and age. The two way interaction between spine level and task was significant (Table 6). There is no effect of ROM for LLUL and ULLT for task, however SLL (M=11.74±3.01) lead to larger ROM during W than SD (M=6.47±1.19) [Figure 10]. Furthermore, there was not a main effect for age on contralateral rotation ROM (*p*=1.00).

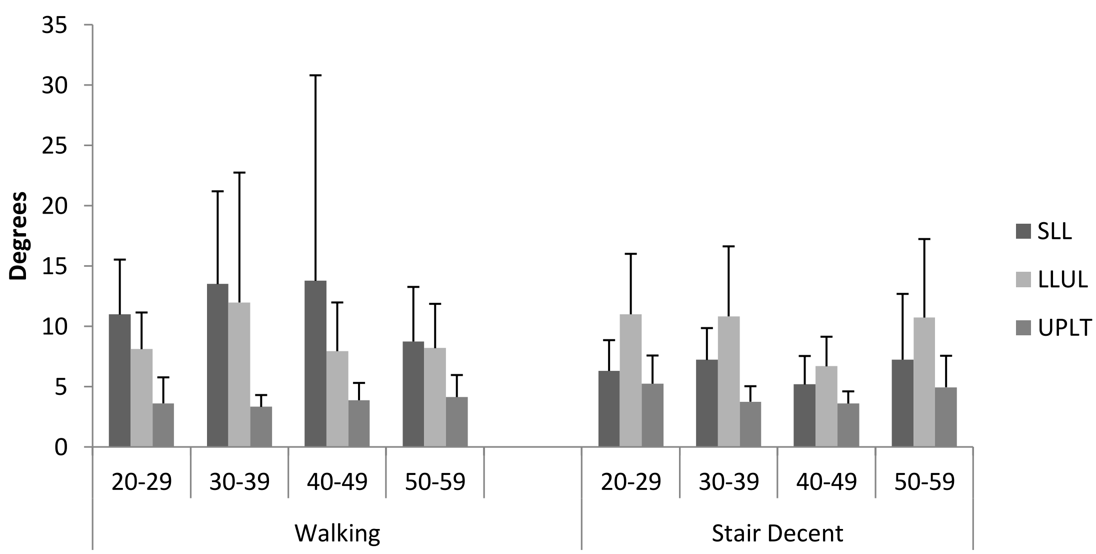


Figure 10. Rotation range of motion; different age groups are listed the horizontal axis. Spine level angles are labeled in solid black for walking task and gray for stair descent.

Table 7. Contralateral rotation maximum peak angle and ranges of motion for multiple spine levels during two ambulatory tasks of daily living.

Spine Level	SLL	LLUL	UPLT
Maximum Peak Angle (deg) Mean (Stdv)			
Task (Walking)	5.99 ± 1.09	3.73 ± 0.81	1.37 ± 0.59
Task (Stair Decent)	2.93 ± 0.88	4.59 ± 0.84	1.82 ± 0.60
Overall	4.53 ± 1.02	4.14 ± 0.82	1.59 ± 0.59
Walking			
Age (20-29)	5.49 ± 1.28	3.03 ± 0.95	1.73 ± 0.57
Age (30-39)	6.95 ± 2.86	4.38 ± 1.27	0.77 ± 1.04
Age (40-49)	8.41 ± 3.29	2.85 ± 1.87	-0.10 ± 1.45
Age (50-59)	3.10 ± 1.22	5.04 ± 2.47	3.08 ± 1.57
Stair Decent			
Age (20-29)	1.54 ± 1.60	4.90 ± 1.21	2.82 ± 0.77
Age (30-39)	5.22 ± 1.76	3.27 ± 2.21	0.96 ± 1.16
Age (40-49)	3.05 ± 1.73	3.42 ± 1.14	0.06 ± 1.49
Age (50-59)	2.54 ± 2.57	7.02 ± 2.75	3.54 ± 1.43
Range of Motion (deg) Mean (Stdv)			
Task (Walking)	11.67 ± 1.70	8.80 ± 0.95	3.75 ± 0.28
Task (Stair Decent)	6.39 ± 0.58	9.74 ± 0.88	4.43 ± 0.34
Overall	9.12 ± 1.36	9.74 ± 0.92	4.08 ± 0.32
Walking			
Age (20-29)	10.97 ± 1.61	8.10 ± 1.08	3.60 ± 0.76
Age (30-39)	13.49 ± 3.15	11.97 ± 4.40	3.31 ± 0.40
Age (40-49)	13.77 ± 6.02	7.93 ± 1.43	3.86 ± 0.51
Age (50-59)	8.73 ± 1.60	8.18 ± 1.30	4.12 ± 0.65
Stair Decent			
Age (20-29)	6.29 ± 0.90	10.99 ± 1.77	5.23 ± 0.83
Age (30-39)	7.20 ± 1.08	10.82 ± 2.37	3.74 ± 0.53
Age (40-49)	5.17 ± 0.84	6.67 ± 0.87	3.59 ± 0.36
Age (50-59)	7.23 ± 1.93	10.72 ± 2.30	4.94 ± 0.92

Discussion

This study examined segmental spine kinematics and its relationship to two ambulatory tasks of daily living in order to demonstrate the effect of aging on spine biomechanics. Age did not influence ROM or peak spinal angles in any of the three planes of motion. Different daily living tasks influenced the sagittal plane ROM where the SD was significantly smaller than the W task. Furthermore, the SD task produced larger contralateral rotation ROM in the LLUL joint than the ULLT joint. The SLL joint ROM in the sagittal plane was larger than ULLT. Additionally, the SLL had a larger ROM than LLUM in the sagittal plane. Spine levels had different ROM in the frontal plane, where the SLL joint had larger ROM than both LLUL and ULLT spinal joints.

The anatomic location of the fifth lumbar vertebra and the unique spatial orientation of the facet articulation with the sacrum may indicate why larger ROM were observed in the SLL joint (White & Panjabi, 1990). In addition to the orientation of the facet joints, the SLL joint has the largest intervertebral disc of the entire spine and White and Panjabi (1990) have shown the size of the intervertebral discs strongly correlates with the ROM available at that joint. In the sagittal plane of motion, different spine levels had different ranges of motion. SLL ROM was significantly larger from ULLT, indicating that some of the forward bending may come from a more inferior area of the spine. Similarly, SLL ROM was larger than LLUM peak angle. These results may also suggest the body may be positioning itself in a way needed to develop momentum to complete the task. This is done by positioning of the whole body COM closer to the anterior boundary of the base of support (BOS). This body orientation will provide a larger forward moment created by the COM and will assist in the forward momentum of the body (D. A. Winter, 2009). The larger ROM at the SLL joint is directly related to the amount of mass above the joint. As the distance from the SLL joint increases superiorly, the amount of mass to aid in momentum generation also increases. Therefore, smaller anterior motion from the SLL joint is needed to generate the required momentum to compete the task. If momentum generation occurred at a more superior joint, there would need to be more anterior lean to produce the require momentum.

Similar to the sagittal plan motion, ranges of motion in the frontal and transverse plane were found to be the largest at SLL. This increased movement in the SLL joint could be a possible mechanism as to why a higher percentage of back pain is reported in the low back compared to the upper back (NINDS, 2011). As increased motion in the spine has been discussed as a mechanisms for back pain and back injury (White & Panjabi, 1990). Increased motion at the spine would suggest larger moment production at the intervertebral discs and increased moments are a risk factor for the causation of back pain (Chaffin et al., 2006). These results could also suggest that the lower spine is responsible to absorbing the lower extremity motion from the ambulatory tasks which allows the upper spine to keep the head oriented forward. Further investigation is needed to better understand this possibility.

With data included in this current study, age was not found to have an overall significant effect on any of the outcome variables. This result is surprising as the age related changes have been reported previously. For example, it has been reported the lower extremity parameters such as knee extension, stride length and peak hip extension are affected by age (Kerrigan, Lee, Collins, Riley, & Lipsitz, 2001; Ostrosky, VanSwearingen, Burdett, & Gee, 1994). Additionally, a systematic decrease in the center of mass (COM) motion with age has also been shown (Hahn & Chou, 2004). Furthermore, a observed increase in the flexion range of motion suggest that as people age, the trunk posture leans more anteriorly as suggested by DeVita and Hortobagyi (2000). Finally, increases in lateral bending in the COM M/L sway have been reported in a aging population (H. J. Lee & Chou, 2006).

Other studies have suggested age does not change trunk kinematics For example, McGibbon and Krebs (2001) were unable to detect age-related differences in absolute trunk and pelvis ROM and peak pitch angles during gait. It was found as the subjects got older (>60 years) they exhibited less low-back (trunk relative to pelvis) range of motion (McGibbon & Krebs, 2001). Additionally, it was found that pelvic rotations in sagittal, frontal and transverse planes of motion were systematically reduced in individuals over sixty years old (Van Emmerik et al., 2005). Furthermore, in subjects over sixty years old it was found that individuals had reduced trunk flexion–extension in the sagittal plane as well as an increased trunk rotation in the transverse plane compared to younger individuals (Krebs et al., 1992; Van Emmerik et al., 2005). This might suggest more conclusive age related segmented spine kinematic effects may be found in individual's great than sixty years old.

In the sagittal and transverse planes the W task had larger peak angles and ROM than SD for several of the spine levels and age groups. This might suggest less momentum generation is needed during the SD task as the SD task is assisted by gravity during the reduction in elevation, where the level walking must continuously generate momentum to assist in the ambulation of the body. Therefore with the requirement of less momentum, less movement is needed to complete the SD task.

A possible limitation to this study was the number of subjects who participated. This small number of participants may have resulted in larger variation of the individual parameters, which may have resulted in the insignificant trends observed in the outcome variables of interest with this method. Although the authors believe this is a reasonable explanation, due to the small excursions of the segmented spine model, previous literature may suggest otherwise. For example, a population of thirty (30) individuals were able to detect a differences in trunk/spine biomechanical parameters (Van Emmerik et al., 2005). A different study had a subject population of ninety-three (93) and were not able to detect differences in truck biomechanical parameters (McGibbon & Krebs, 2001). This might suggest that other factors may be contributing to the unobserved age differences.

Another possible limitation was the ages of the individuals in the study. The oldest individual was less than sixty (60) years old, and the subjects simply may not have started to present with the related effects of aging. The lower extremity aging studies generally look at individuals as old as seventy or seventy-five years old (Hahn & Chou, 2004; JudgeRoy et al., 1996; McGibbon & Krebs, 2001; Van Emmerik et al., 2005). It would be most advantageous to include a subject group or groups up to seventy (70) or eighty (80) years of age. Although this would be most interesting experimentally, this is a very frail population and the ability to test such a cohort of people may be limiting.

In conclusion, this investigation applied an in-vivo segmental spine approach to determine if age related changes in certain biomechanical parameters (maximum peak angle and range of motion) could be observed during two different tasks. Task was found to have some influence over the observable spine angles and ROM, while age was found to not have any influence over the outcome measures. A possible reason for this is the age of the testing population was possibly not old enough and had not fully exhibited the effects of aging. It is suggested that with a higher number of subjects and an increase in the age of the subjects, an age related difference in spine kinematics may present itself. Thus further study on a healthy population is needed before applying this protocol to a pathologic cohort. Although the hypothesis was not fully supported differences were shown to exist in abundant between segmental spine angle levels. This is encouraging that this method of segmented spinal motion can continue to be used to investigate and understand the complexity of the spine and the motions it produces.

Bridge

. Chapter IV examines the relationship between segmental spinal angles, age and different ambulatory task. Task was shown to have some influence on the different segmental spinal angles, while age was found to have no influence on the segmental spine biomechanical. This was possibly due to the age of the individuals and it was suggested the same protocol should be applied to individuals who are even older as reported in previous literature. In chapter V, the reaction forces of the segmental spine will be investigated and how various ambulatory tasks of daily living may affect these reaction forces.

CHAPTER V

SPINE JOINT REACTION FORCES DURING MULTIPLE ACTIVITIES OF DAILY LIVING

This chapter was developed by Li-Shan Chou, Ph.D. & Scott P. Breloff. Dr. Chou contributed substantially to this work participating in the development of methodologies and providing invaluable critiques and substantial editing advice. Scott P. Breloff was the primary contributor to the development of the protocol, data collection, data analysis and did the writing.

Introduction

There are many different mechanisms thought to cause back pain, (M. A. Adams, Freeman, Morrison, Nelson, & Dolan, 2000), including nerve or muscle irritation, bone lesions, degeneration in the intervertebral disc from acute and repetitive loading (M. Adams & Hutton, 1981; Goel & Weinstein, 1990; Nachemson & Elfstrom, 1970; NINDS, 2011). Additionally, a correlation has been shown between the mechanical loading in the back and the presence of back pain (Wilke, Neef, Hinz, Seidel, & Claes, 2001). Therefore a better understanding of the mechanical loadings, such as joint reaction forces, in the lumbar and thoracic spine, are important in determining if intervertebral disc degeneration is associated with back pain (Nachemson & Elfstrom, 1970). The quantitative methods that are used to describe the forces in the spine are either invasive or estimated only at the 5th lumbar 1st sacral joint (Ledet, Tymeson, DiRisio, Cohen, & Uhl, 2005; Park & Chaffin, 1974). It is the expectation that results from this study could be used to better understand possible biomechanical mechanisms which could lead to back pain, thereby aiding in the development of better treatment methods and new preventative procedures.

Simplistic lifting models have been used to estimate the load exerted at the 5th lumbar/1st sacral (L5/S1) or 4th lumbar/5th lumbar (L4/L5) joint in both static and dynamic situations (Freivalds et al., 1984; Hall, 1985; Hwang & Kim, 2009; McGill & Norman, 1986; Park & Chaffin, 1974). Due to the intrinsic complexity of the spine, simplistic spine models are not able to fully estimate changes to the spine due to various dynamic perturbations. Despite these limitations, simplistic spine models are commonly used, even though they neglect the intricacy of the seventeen bones and over thirty joints in the lumbar and thoracic spine. Although the study of lifting is important, many people do not spend most of their time performing this task. Instead, more details should be applied to how forces in a segmented spine respond during ambulatory activities of daily living (ADLs), such as walking (W), obstacle crossing (OC), stair ascent (SA) and stair descent (SD).

Estimating the joint reaction forces in the spine during dynamic tasks has been reported previously using simple surface marker models. Khoo et al. (1995) developed a method to calculate the forces in the lumbosacral joint, where the spine was included with the head arms trunk (HAT) during normal walking. Furthermore, estimations of lumbosacral forces have been investigated during backpack loads and reported that lumbosacral force can increase when carrying a backpack (Goh et al., 1998; Hong & Cheung, 2003). Additionally, joint reaction forces at the L4/L5 joint have been reported during normal over ground walking during varying (fast, normal and slow) speeds and found forces well over the body weight (Callaghan et al., 1999).

Modeling the spine with a large segment does not seem the most accurate or most informative approach, as pressure between individual vertebra has been previously reported by inserting a needle directly into intervertebral discs in humans (Polga et al., 2004). Nachemson and Elfstrom (1970) also used a needle directly implanted into the intervertebral (between L3 & L4) disc to measure force where intradisc pressures were reported for various tasks activities of daily living such as walking, coughing, straining, laughing, etcetera. Additionally, two patients were implanted with telemeterized spinal fixators (spanned L3 & L4, respectively) and data were collected during various tasks including walking, stair ascent and descent and resultant forces were not found to be different between walking and stair negotiation tasks (Rohlmann et al., 1997). Animal models have been used in lieu of humans due to the invasiveness of this procedure. Some direct measurements were made in baboons during dynamic movements and reported L4/L5 forces between two and four times body weight (Ledet, Sachs, Brunski, Gatto, & Donzelli, 2000; Ledet et al., 2005).

Although direct measurements of intervertebral discs are the most accurate to fully understand the pressures and forces associated with motions in the spine, these procedures are very invasive and would be difficult to apply to a large clinical population. Therefore, the development of a non-invasive but detailed method to estimate the reaction forces in the spine will allow for a more complete understanding of the forces associated with the spine during ambulatory activities in a large number of individuals.

The development of an in-vivo segmental spine procedure which will allow for detailed quantification of spinal joint reaction forces during various tasks is important. A process like this will allow for a more complete understanding of the reaction forces at the spine during ambulatory activities. The purpose of this study is to explore the feasibility of an in-vivo multi-segment spine marker set used to quantify the joint reaction forces at various spinal joints during different activities of daily living in young adults. It is hypothesized that unique ambulatory activities of daily living will affect joint reaction forces at specific joints of the spine. Specifically, it is thought that obstacle crossing and stair ascent will be larger than walking due to the increased force needed to negotiate the obstacle or increase elevation of the stairs. Additionally, it is thought stair descent will be less than walking, as gravity will assistant in the decrease in elevation.

Methods

Subjects

Fourteen healthy young adults (7 males/7 females; mean age: 27.9±5.9 years, mean height 176.0±27.7cm, mean mass 67.8±17.2 kg), were recruited from the university community to participate in the study. Subjects did not have a history or clinical evidence of neurological, musculoskeletal or other medical conditions affecting gait performance, such as stroke, head trauma, neurological disease (i.e. Parkinson's, diabetic neuropathy), visual impairment uncorrectable by lenses and dementia. All subjects reviewed and signed an informed consent approved by the Institutional Review Board.

Experimental Protocol

Male subjects were asked to wear spandex shorts with no shirt. Female subjects wore a dance leotard with open back. They then performed four different bare foot tasks including: level ground walking (W), obstacle crossing (OC), stair ascent (SA) and stair descent (SD). The task order was randomly selected for each subject. The level walking task required subjects to walk along a 10 meter long walkway. For obstacle crossing task, subjects were asked to initiate walking from a distance which allowed at least 3 steps prior to encountering the obstacle, step over the obstacle, and continue walking. The obstacle was set at 10% of body height and made of a polyvinyl chloride (PVC) pipe measuring 1.5 m long and a diameter of 2.5 cm, which was presented to the subjects prior to obstacle crossing trials (Hahn & Chou, 2004). During the SA, subjects were asked to approach the stairs while walking on level ground, ascend the stairs, and continue walking to the end of the elevated walkway. The starting position for each subject was adjusted to allow at least three steps before stepping onto the first stair. Subjects initiated their SD trials from the back end of the elevated walkway, descended the stairs, and continued walking for several steps.

Whole body motion analysis was performed with a ten-camera motion analysis system (Motion Analysis Corporation, Santa Rosa, CA). Sixty-two retro-reflective markers (diameter=14mm) were placed on the subject. In addition to a whole body marker set (Hahn and Chou, 2004), eight markers were placed directly on the palpated spinous processes of the following vertebrae; Cervical 7 (C7), Thoracic 3 (T3), Thoracic 6 (T6), Thoracic 9 (T9), Thoracic 12 (T12), Lumbar 3 (L3), Sacrum 1 (S1) & Sacrum 5 (S5). Two makers were placed on the left and right posterior superior iliac spine (PSIS), and the remaining markers were placed 50mm to the left and right of the spinous process markers (except for S1 and S5). Three-dimensional marker position data were collected at 60 Hz and low-pass filtered using a 4^{th} order Butterworth filter with the cutoff frequency set at 5 Hz.

Identical force plate configurations were used for the non stair related tasks [Walking (W) & Obstacle Crossing (OC)]. Three force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were placed in series and embedded level into the laboratory floor. The first two force plates were immediately adjacent to one another, and the third plate was separated by a distance of 15cm. This setup was to accommodate subjects walking with different step lengths. Gait events such as heel strike (HS) and toe off (TO) were detected using the vertical ground reaction force (GRFv). Heel strike was determined to occur when the GRFv was greater than 10% of the maximum GRFv, and toe off was determined to occur when the GRFv was less than 10% of the maximum GRFv (Ghoussayni et al., 2004; Hreljac & Marshall, 2000; Mickelborough et al., 2000).

For stair ascending (SA) and descending (SD) tasks, a staircase included three steps was used (Figure 1). Each step had a rise of 17.8 cm, a run of 30.5 cm and a width of 80 cm, forming a stair angle (rise/run) of 30^0 (H. J. Lee & Chou, 2006) A total of four force plates (Advanced Mechanical Technologies, Inc., Watertown, MA) were used to obtain ground reaction force data during SA and SD trials. Two force plates were embedded level into the laboratory floor and two made up the steps (see Figure 2, Chapter III).

Kinematic Analysis

Marker position data were analyzed for one activity cycle of each condition. Five trials were captured for each condition. A gait cycle during level ground walking was defined as the time interval between two consecutive ipsilateral heel strikes (FP 3 to FP 1; see Figure 3a, Chapter III). The obstacle crossing stride was defined as the heel-strike of the leading limb before the obstacle to the heel-strike of the same limb after clearing the obstacle (FP 3 to FP 1; see Figure 3b, Chapter III). Stair ascent was examined for the duration between consecutive ipsilateral heel strikes of last level ground contact and the second stair (FP 2 to FP 4; see Figure 3c, Chapter III), and stair descent was examined consecutive ipsilateral heel strikes following first step down to ground (FP 4 to FP 2; see Figure 3d, Chapter III).

A MATLAB® (Mathworks, Natick, MA) program was developed to calculate six adjacent segmental spinal forces. The joints in the current study are labeled as: sacrum to lower lumbar [SLL], lower lumbar to upper lumbar [LLUL], upper lumbar to lower thorax [ULLT], lower thorax to lower middle thorax [LTLMT], lower middle thorax to upper middle thorax [LMTUMT] and upper middle thorax to upper thorax [UMTUT] (see Figure 1, Chapter III).

Kinetic Analysis

Segmented spine joint reaction forces were calculated using the Inverse Dynamics algorithm through a rigid body linkage starting with the lower extremities. Lower extremity forces are transferred to the spine through the hip joints and pelvic segment, which includes the femur heads, where hip joint reaction forces are well documented, and the 1st sacral vertebra (D. A. Winter, 2009). The joint between the ilium and sacrum has an irregular surface, thus little or no force is lost across that joint (Hoek van Dijke, J Snijders, Stoeckart, & Stam, 1999). This characteristic was paramount for the single segment definition of the pelvis (Figure 4). The center of mass (COM) of the pelvis was defined as the mid-point between the mid-point of the left and right anterior superior iliac spine and the mid-point between the left and right posterior superior iliac spine. The mass of the pelvis COM was calculated using regression equations (Dempster, Gabel, & Felts, 1959). The pelvis COM segmental acceleration was calculated using the procedure described in D. A. Winter (2009). To do this, first time-varying rotation angles were calculated prior to transforming from the global coordinate system to the anatomical coordinate system. Next the first derivative of these angles provided the components for the segmental angular velocities. The angular velocity of the first rotation was set equal to the first time-varying rotation angle. Then the second time-varying rotation angle is added to the matrix product of the first angular velocity and the rotation matrix for the second angular velocity. This is repeated for the third segmental angular velocity. Finally, the third angular velocity matrix is decomposed along the three anatomical axes to allow time-varying three dimensional segment angular velocities (Figure 4). The hip reaction forces, pelvis segmental mass and pelvis segmental acceleration were summed to provide the reaction force at the 1st sacral joint [equation 1 & 2] (D. A. Winter, 2009).

$$\overrightarrow{Jrf_{S1}} = m\vec{a}_{pelvisCOM} - m\vec{g}_{pelvis} - \vec{f}_{r.hip} - \vec{f}_{l.hip}$$

Equation 1. Vector form equation for reaction force at sacral joint.

$$\overrightarrow{Jrf}_{proximal\ spine\ segment} = m\vec{a}_{pelvisCOM} - m\vec{g}_{pelvis} - \vec{f}_{distal\ spine\ segment}$$

Equation 2. Vector form equation for reaction force at subsequent spine segments.

Segmental accelerations were determined at the COM of each spine segment using the procedure outlined (D. A. Winter, 2009). The mass of each spinal segment was calculated by first modeling the trunk segment as homogeneous rigid bodies. Then, several virtual markers were added in front of the trunk and estimated by eight solid ellipsoids. The trunk shape was described by 1027 tetrahedrons defined by the surface and virtual markers (Figure 1). The volume of each ellipsoid was then calculated using both actual and virtual markers. The density of the human body was represented by 1.063 kg/cm^3 (Krzywicki & Chinn, 1967). The density of the human body was then multiplied by the volume of each segment to determine the mass (Figure 2).

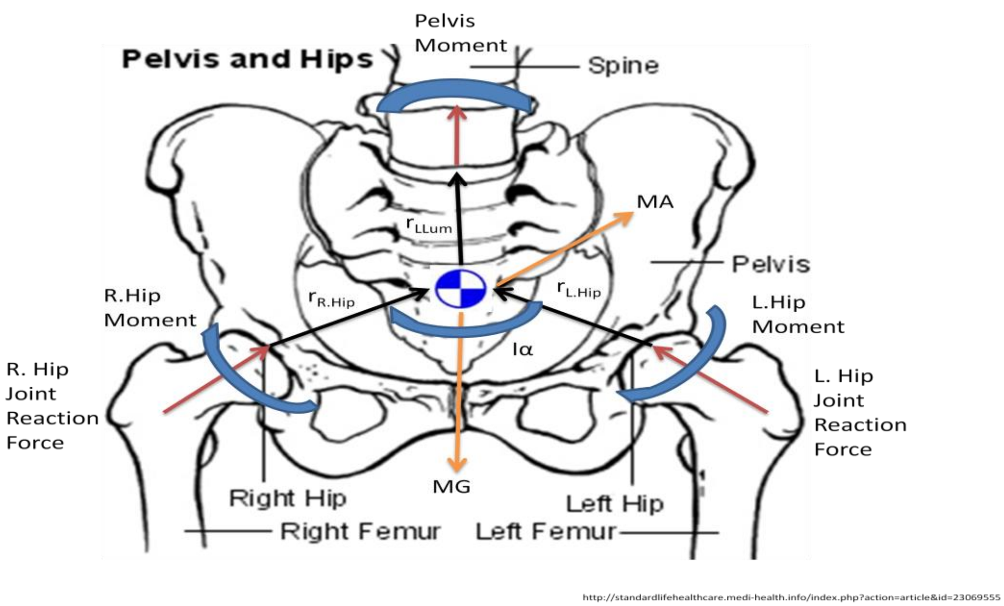


Figure 1. Free body diagram of the pelvis. Showing how lower extremity forces will be handled to continue the summation of forces into the spine.

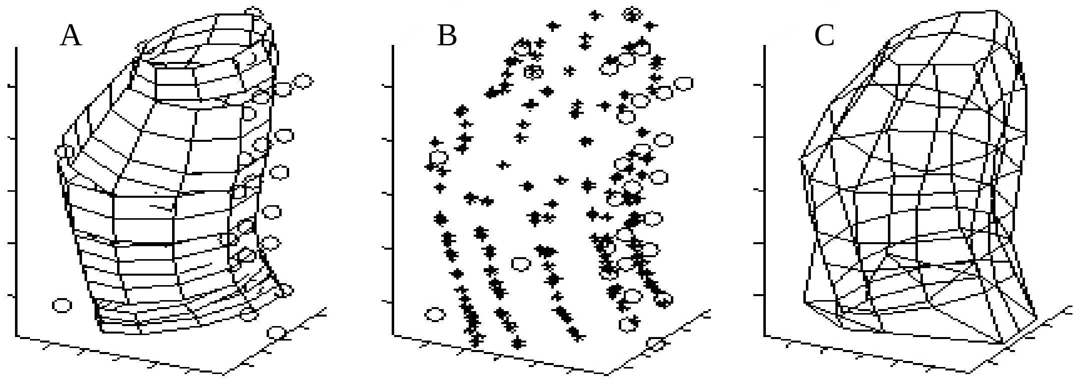


Figure 2. (A) Eight ellipsoids indicated by the attached markers (circles). (B) Virtual markers (asterisks) estimated by the ellipsoids. (C) The trunk shape described by tetrahedrons.

Validation

To validate the current procedure of spine joint reaction force calculation, lower extremity joint reaction forces were compared between the current method and previously validated third party software (Orthotrac software; Motion Analysis Corporation, Santa Rosa, CA). The two joint reaction force calculation methods (MATLAB & Orthotrac) produced similar wave forms, which were then evaluated using the CMC (Ferrari et al., 2010; Kadaba et al., 1990). Data were found to have high correlations (>0.90), which is accepted as high in all directions. These results suggested the calculation procedures between MATLAB and OT were similar (Ferrari et al., 2010; Kadaba et al., 1990). Therefore, the results observed from the segmented spine can be considered reasonable based on procedure. An example of the vertical reaction forces are shown below, though all directions had similar comparisons between the two methods (Figure 3).

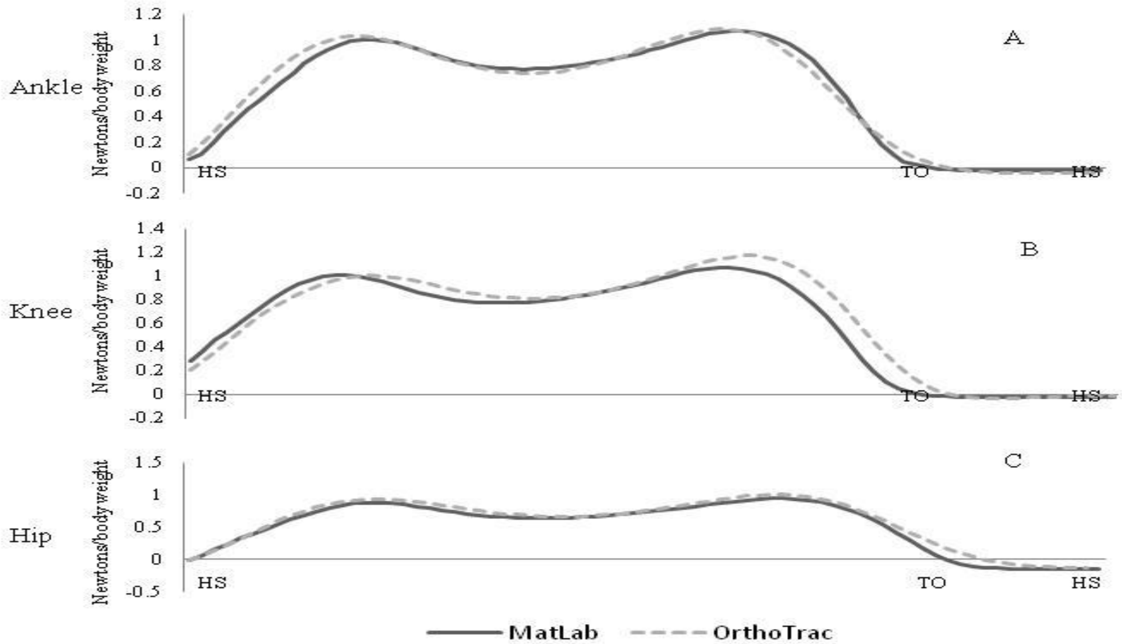


Figure 3. Ensemble average lower extremity vertical joint reaction forces as calculated by OrthroTrac and MATLAB; A) Ankle, B) Knee & C) Hip.

It has been shown reaction forces have similar shape moving from distal to proximal with smaller magnitudes (D. A. Winter, 2009). To ensure this characteristic was present in the segmented spine, static trial reaction forces were calculated at each spine segment. The results were exactly as expected with the most distal segment (Sacrum to Lower Lumbar) displaying the largest magnitude with each proximal segment yielding slightly less force production (Figure 4).

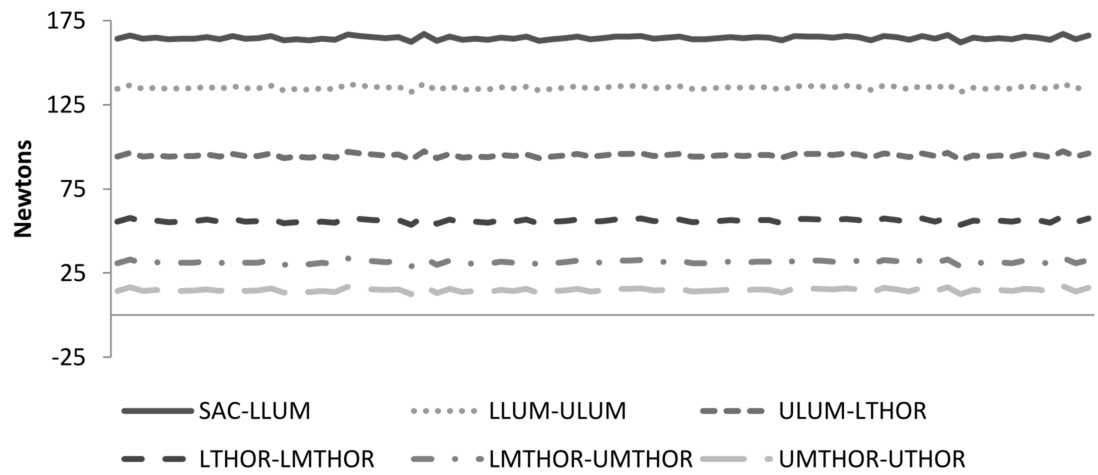


Figure 4. Static joint reaction forces at each spine level.

Data were analyzed using a two-way within factor analysis of variance. Data were normalized by body mass for statistical comparison between individuals. The dependent variable was the estimated peak spine reaction force. Task was a within subject effect with four levels: (a) level walking, (b) obstacle crossing, (c) stair ascent, and (d) stair descent. The second factor, also a within subject effect, was spine level with six levels: (a) sacrum to lower lumbar [SLL], (b) lower lumbar to upper lumbar [LLUL], (c) upper lumbar to lower thorax [ULLT], (d) lower thorax to lower middle thorax [LTLMT], (e) lower middle thorax to upper middle thorax [LMTUMT] and (f) upper middle thorax to upper thorax [UMTUT]. In all cases, adjusted p-values (Greenhouse-Geisser) were used to evaluate within subject effects because the assumption of sphericity was evaluated with the Mauchly Sphericity Test and found to be non-tenable, $p < 0.05$.

Results

A visual inspection of the each segmental joint reaction force found each spine level to have very similar patterns, therefore the ensemble average of all subjects for one joint segment from each task is shown (Figure 5). Data were represented visually in a non-normalized manner for illustration purposes only. Vertical segmented spine joint reaction forces had the highest magnitude compared to the other two planes of motion. Additionally, the vertical joint reaction force is bimodal, similar to the lower extremity vertical ground reaction force curve during walking (Figure 5A). Medial-lateral segmented spine joint reaction forces were found to have the least amount of spinal reaction force. Stair ascent appeared to have a larger shear force than the other tasks, however it was not significantly different (Figure 5B; Table 1). Anterior-posterior segmented spine joint reaction forces were found to have similar patterns between walking, obstacle crossing and stair descent and they appeared to be bimodal. Stair ascent, presented an anterior-posterior segmental shear force that was generally larger than the other tasks and did not have a bimodal pattern (Figure 5C).

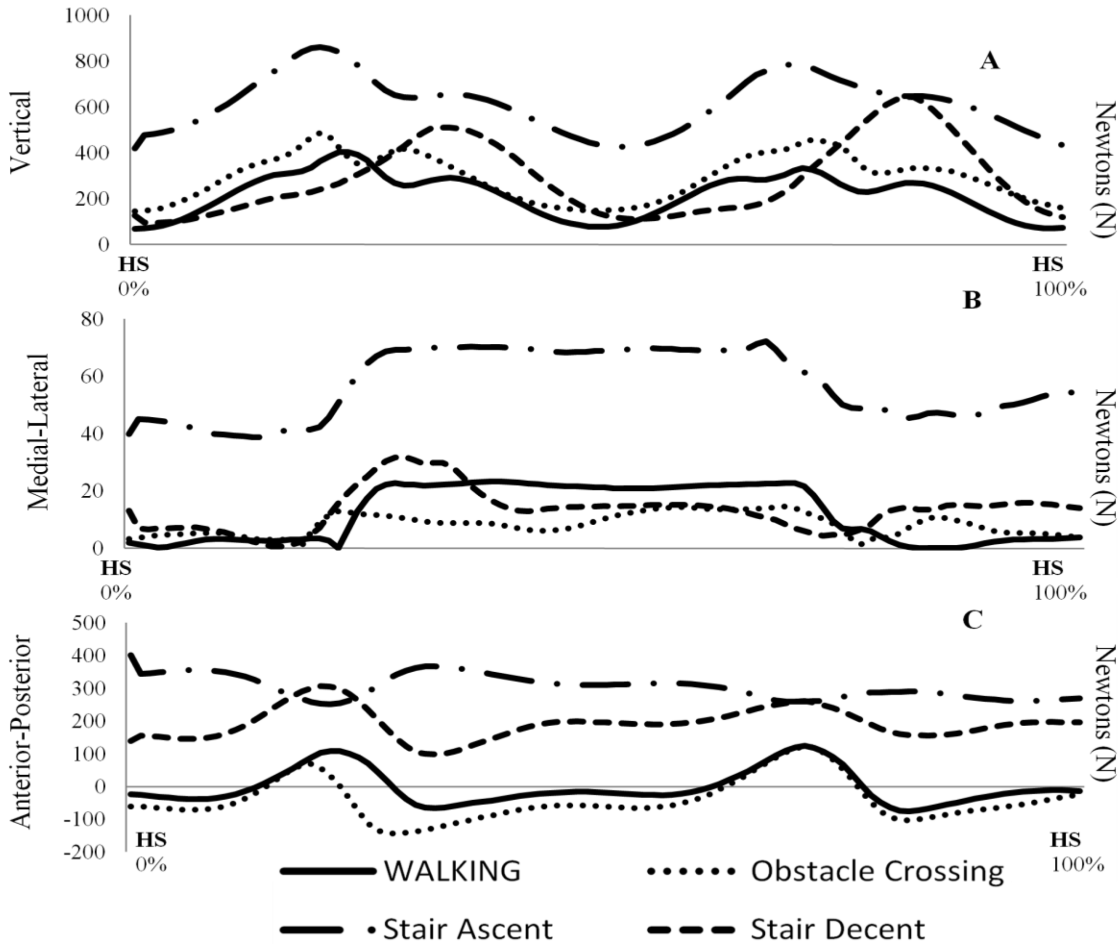


Figure 5. Lower thorax to lower middle thorax (LTLMT) ensemble average segmented spinal joint reaction forces for each plan of motion during each of the four ambulatory tasks of daily living. Visual inspection found all spine levels to have similar patterns, thus the LTLMT was chosen as a representative sample.

Anterior Posterior Segmental Force Peaks

A non-significant interaction was found between task and spine level for the anterior-posterior segmental force peaks (p = 0.429). Main effects for task (p = 0.628) and spine level (p = 0.952) were not (Figure 6).

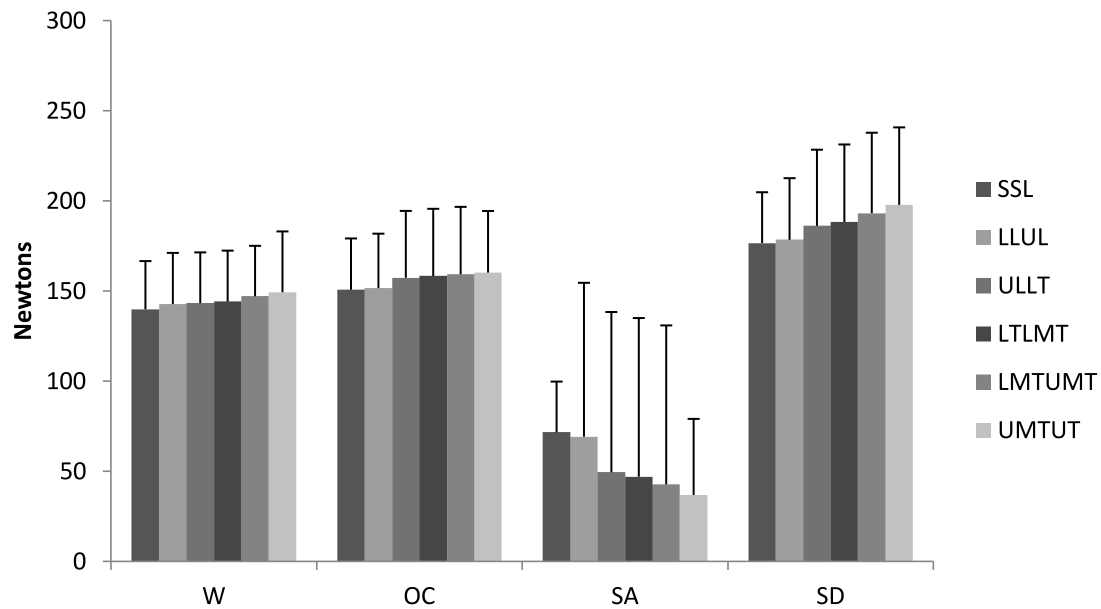


Figure 6. Anterior/posterior non-normalized spine joint reaction forces for spine level and multiple tasks.

Medial Lateral Segmental Force Peaks

The interaction was found to be non-significant for the peak segmental forces in the medial-lateral direction (p =0 .999). The main effects for both task (p =0 .536) and spine level (p =0 .772) in relation to the medial-lateral force peaks were not-significant (Table1; Figure 7).

Table 1. Non-normalized Maximum peak segmental joint reaction forces in the anterior/posterior, medial/lateral and vertical directions for each activity of daily living.

	W	OC	SA	SD	p-value (main effect) -normalized-
Maximum Anterior/Posterior Joint Reaction Force (N) Mean (Stdv)					
Task (Overall)					.628
Spine Level (Overall)					.952
SSL	139.88 ± 26.71	150.74 ± 30.05	71.61 ± 85.70	176.61 ± 33.75	
LLUL	142.72 ± 28.41	151.64 ± 30.16	69.13 ± 85.41	178.43 ± 34.16	
ULLT	143.29 ± 28.13	157.36 ± 37.07	49.48 ± 88.86	186.16 ± 42.22	
LTLMT	144.21 ± 28.18	158.31 ± 37.27	135.21 ± 243.21	188.29 ± 43.00	
LMTUMT	147.05 ± 28.01	159.27 ± 37.40	42.75 ± 88.16	193.15 ± 44.64	
UMTUT	149.31 ± 28.55	160.22 ± 37.74	36.82 ± 87.46	197.78 ± 46.72	
Maximum Medial/Lateral Joint Reaction Force (N) Mean (Stdv)					
Task (Overall)					.536
Spine Level (Overall)					.772
SSL	49.49 ± 11.69	45.86 ± 14.46	34.62 ± 6.82	51.00 ± 21.35	
LLUL	43.75 ± 12.71	45.51 ± 14.00	34.38 ± 6.84	51.39 ± 21.48	
ULLT	43.13 ± 13.02	46.09 ± 13.58	34.08 ± 6.98	53.75 ± 20.83	
LTLMT	42.60 ± 13.48	45.71 ± 13.45	34.36 ± 7.04	54.20 ± 20.97	
LMTUMT	42.41 ± 13.80	45.52 ± 13.47	34.51 ± 7.89	54.33 ± 20.99	
UMTUT	42.34 ± 14.00	45.46 ± 13.49	40.47 ± 17.65	54.34 ± 21.04	
Maximum Vertical Joint Reaction Force (N) Mean (Stdv)					
Task (Overall)					<.001
Spine Level (Overall)					.547
SSL	621.30 ± 232.98	719.94 ± 237.83	654.34 ± 206.41	880.44 ± 341.95	
LLUL	583.31 ± 223.96	681.14 ± 226.99	651.57 ± 157.85	844.58 ± 332.94	
ULLT	529.98 ± 212.83	625.63 ± 222.72	608.51 ± 118.51	801.12 ± 324.41	
LTLMT	499.55 ± 205.78	594.36 ± 213.38	586.87 ± 108.33	772.89 ± 316.53	
LMTUMT	474.58 ± 200.92	568.63 ± 207.30	566.72 ± 100.45	748.88 ± 312.91	
UMTUT	456.23 ± 196.54	549.40 ± 202.05	552.02 ± 96.32	731.64 ± 310.87	

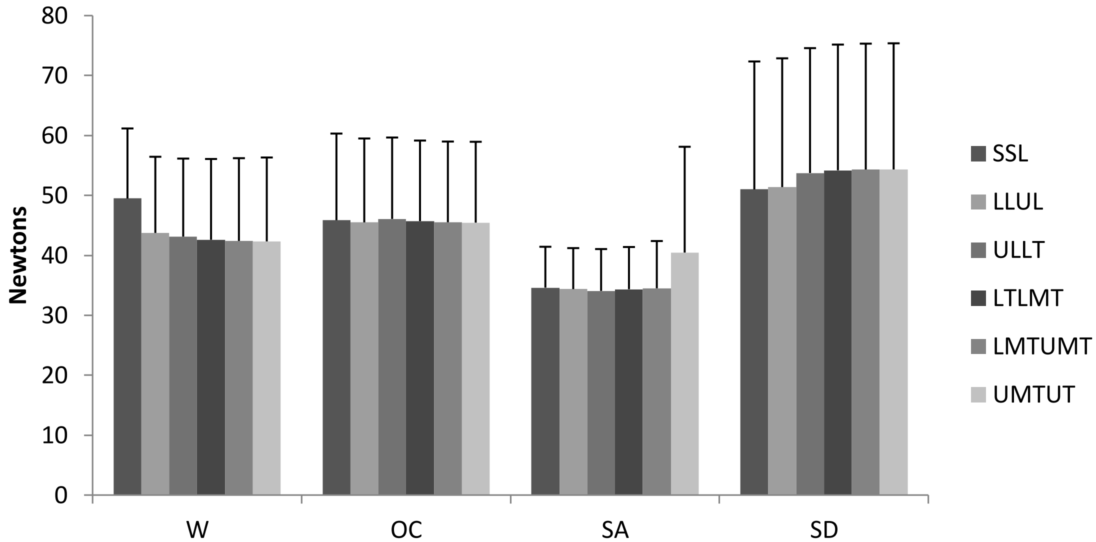


Figure 7. Medial/Lateral non-normalized peak spine joint reaction forces for spine level and multiple tasks.

Vertical Segmental Force Peaks

The two way interaction for vertical segmental force peaks of task and spine level was not significant (p = 0.842). A significant main effect was found for task (p < 0.001) [Table 1]. Post-hoc pairwise comparisons of the task marginal means showed the W (M=8.05±3.19 N/kg) task had significantly smaller peak reaction forces than the SD (M=12.12±1.32 N/kg) task (p = 0.007). Additionally, it should be noted the OC (M=9.43±0.76 N/kg) task was trending to have larger reaction forces than W, (p = 0.009). The spine level main effect (p = 0.547) was non-significant (Figure 8).

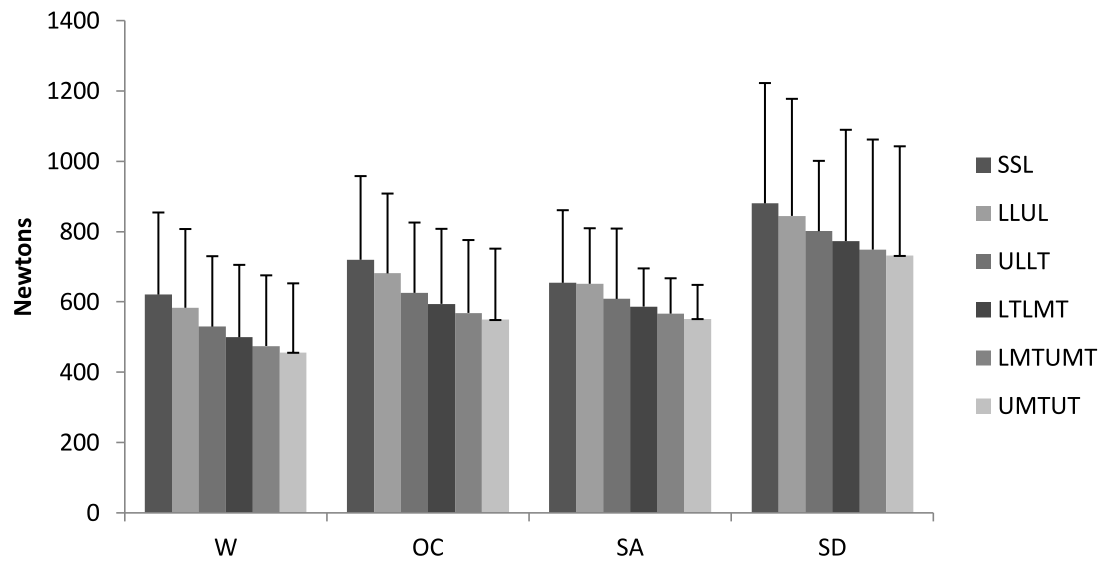


Figure 8. Vertical peak non-normalized spine joint reaction forces for spine level and multiple tasks.

Discussion

This study sought to compare peak spine joint reaction forces of a multi-segmented spine marker set during four distinct ambulatory activities of daily living. All of the interaction tests were non-significant, therefore only a general discussion relating how the factor (task or spine level) would influence the observed peak force is presented.

Task was found to have an overall influence on the peak segmental joint reaction forces in the vertical plane of motion (Table 1). Only one individual main effect differences could not be detected due to the Bonferroni correction factor. It is encouraging to find the segmental spine approach has the ability to detect general differences between tasks in at least one plane of motion.

Although main effect differences were only observed between tasks for the vertical peak reaction forces this could be attributed the similarity in the speed of the tasks (Figure 10). All tasks were completed at the subjects own comfortable pace, which generally indicates an average walking speed of approximately 1.50 m/s for young adults (Carey, 2005). It has been shown a slow jog, or speeds greater than 2.0 m/s are required before changes in vertical ground reaction forces are observed (Keller et al., 1996). Thus the self selected speeds of the individuals in this study might not have induced changes in the ground reaction force which in turn would have induced a change in the peak spine joint reaction forces. Therefore, future studies could be conducted while using varying speeds to determine if there is an observable difference in spinal joint reaction forces as speed increases. This has some application in daily life and work place application; however results from a speed study would have a much larger sports application.

Segmental spinal joint reaction forces in all three planes of motion were not different between spine segments (Table 1). A possible explanation for this could be due to the tasks used in the current study. These tasks were primarily gait tasks (walking, obstacle crossing, stair ascent and descent), which require the subjects to remain mostly vertical, and not have much inter-trunk movement. Due to the lack of overall inter-trunk movement, the accelerations by the individual spine segments would be similar. Therefore the joint reaction forces between segments with similar mass and comparable accelerations will generally not be statistically significantly different. Therefore, this procedure may be very useful for movements which require inter-trunk movements such as lifting or sports applications.

One limitation of this study was the lack of electromyography (EMG) measures of the trunk muscles. The spine joint reaction force is comprised of force contact between the intervertebral discs and vertebral bodies and all the muscles forces around the spine.

An EMG recording of the trunk muscles (i.e. erector spinae) during the various task of daily living might enhance the understanding of the peak spine joint reaction forces. If muscle activation varies (increases/decreases) with different ADL's then muscle activity must be included into the joint reaction force calculation. If muscle activation does not vary, then the speed of the task would have be the majority contributor to the spine joint reaction force calculation. Another possible limitation to this study was the gait speed was not controlled. Considering gait speed could have a major influence on the joint reaction force, future studies might consider controlling gait speed between subjects and activities of daily living.

Furthermore, although this study only examined spinal joint reaction forces future studies should examine multi-segmented moments as well. For example, significant differences in lower extremity moments have been reported during stair climbing when compared to walking (Nadeau, McFadyen, & Malouin, 2003). Additionally, joint moments at the L4/L5 level have found consistent patterns and different peak moments as walking speed increased (Callaghan et al., 1999). This may suggest that observing segmental spine motion could find that various spine levels produce multiple moments in response to different tasks. Although not a direct application of this study, it is suggested this could be investigated in the future.

Overall, this procedure appears to be effective in estimating the joint reaction forces using a segmental spine model. Although the hypothesis was not fully supported, the results found the main effect of peak reactions forces in the segmental spine can be influence by task. This was a new observation from previous studies and is therefore encouraging. Further validation is important with the inclusion of a larger and more diverse sample set and should include testing with multiple speeds. One future direction for application of this model would to investigate sports applications which involve more inter-trunk motion than walking based tasks.

CHAPTER VI

SUMMARY OF FINDINGS AND CONCLUSIONS

Back pain affects all ages of individuals and drastically affects their daily lives. To better address the back pain issue a full and comprehensive understanding of spine dynamics during activities of daily living is needed. Therefore, the purposes of this dissertation were to develop and validate an in-vivo marker-based motion analysis protocol which could provide reliable dynamic (kinematic and kinetic) quantification of different spinal segments during locomotion. Additionally, follow up investigations were undertaken to 1) examine the effect of different activities of daily living, e.g., level walking, obstacle crossing, stair ascent and descent, on kinematics of individuals spinal segments in young adults; 2) investigate whether individuals in different age groups (20-29, 30-39, 40-49, 50-59 years) would display altered spinal kinematics during level walking; and 3) explore the feasibility to quantify the joint reaction forces at various spinal joints during different activities of daily living in young adults. Overall, it was hypothesized there would be would be observable differences between select biomechanical parameters (ROM, peak angle and peak joint reaction force) for the multi-segmented spine during walking tasks.

Major Findings

The first study presented a new protocol for measuring spinal motion using multiple spine segments. Our results indicated the use of small segments in the spine, consisting of three vertebrae each, can yield different motion patterns at multiple levels of the spine during level walking. Additionally, during walking the multi-segmented ROM values in all three planes of motions had p-values (*p*>0.05) suggesting the ROM values were the same between days. Furthermore, spinal multi-segmented CMC values ranged from ~0.48 to ~0.95 for within day and ~0.28 to ~0.60 for between days, indicating during level walking, the calculated spinal motion patterns were indeed repeatable. Finally, the ICC values assessed the reproducibility/reliability of this new multi-segmented protocol. It was suggested ICC was not the best calculation for reproducibility/reliability of this new marker set due to the sometimes high standard deviations which can be found in these angles. It was decided the repeatability found by the CMC and ROM measures provided the necessary rationale of this procedure.

The second study described how different activities of daily living may influence the observed spinal motion. Results indicated that some differences in peak displacement and overall range of motion exist in the frontal, sagittal and transverse planes. Particularly, during the SA task ULLT frontal plane joint motion had a greater ROM than the W task. Additionally, the SA task also had more ROM than the SD at the ULLT frontal spine level. In the transverse plane of motion, the SLL spine level presented with significantly larger OC ROM than the SD task. Furthermore, SLL joint had the largest ROM in the sagittal plane of motion. It is speculated this is due to the orientation of the facet articulation with the sacrum and the need for the body to develop momentum needed to complete the ADL task.

The third study discussed how the effects of aging influenced spinal motion. The results from this study suggest that there were no age effect. It was suggested this was due to the ages of the subjects and the subjects themselves were still too young to exhibit true aging characteristics. Thus it was suggested that further validation of this model's ability to detect an age effect be pursed further using an older cohort of subjects. Should further study show differences in spine motion due to age, the application of this procedure could be used to develop treatments for aging individuals and help reduce age influenced back pain.

The fourth and final study discussed the expansion of the original kinematic model to include kinetics and strove to determine if the expanded model could detect differences between various daily living tasks. This was done by developing a method of kinetic analysis which allows for the transfer of calculated joint reaction forces from the lower extremity through the pelvis to the spine, where segmental spine joint reaction forces could then be calculated. Task was found to have an overall influence on the peak segmental joint reaction forces in the vertical plane of motion. It is encouraging to find the segmental spine approach has the ability to detect general differences between tasks in at least one plane of motion. It was suggested this was due to the speeds of the various tasks were to close therefore no differences in the ground reactions would be observed thus no difference in the spine reaction forces. Further study may be needed to determine if tasks can effect moment production in spine as it has been shown to do so in the lower extremity.

Limitations of the Study

To our knowledge this was the first study to investigate detailed spinal motion on multiple spine regions (sacrum, lumbar & thoracic) using surface motion markers during various activities of daily living. As with most new protocols we encountered a few limitations with this study.

The first concern regarding this set of experiments is the lack of a 'gold standard' to compare the observed kinematic and kinetic data. Ideally, a gold standard would be three-dimensional fluoroscopy to accurately measure bone movement during the different task. Currently, fluoroscopy is still and expensive and a limited access tool for three-dimensional human motion. The field of view is much too small to record motion from the entire spine, and there is some debate regarding the amount of radiation exposure from this technique in such a sensitive area. Instead the procedure was tested for its repeatability and reliability. The repeatability measure allowed the assurance that the data would be the same from day to day and thus comparable between people.

A second source of potential error could be due to the large inter-subject variability of the multi-segmented spine biomechanical outcome parameters. The sample size may have not been sufficient to fully observe differences in segmented spine motion during ambulatory activities of daily living. Nevertheless, the data was able to direct multiple motion patterns for different activities of daily living. Further studies might consider the incorporation of erector spinae activity during investigations of detailed spine motion.

A third limitation may have been the palpation of the spinous processes during marker placement. Locating the spinous process for marker placement is not as straight forward as in the lower extremity. Several strategies were undertaken to limit this potential source of error. First, only one researcher placed the markers on the spine for every subject, thus eliminating any inter-rater interpretations of what marker placement was correct. Certain more palpable landmarks (e.g. inferior angle of the scapula and scapula spine) were used as reference to allow for a more accurate placement of the markers. In addition, studies which used imaging techniques to accurately determine where the spinous processes location were referenced to ensure accurate marker placement (Harlick, Milosavljevic, & Milburn, 2007).

Another possible limitation was the ages of the individuals in the study. The oldest individual was less than sixty (60) years old, and the subjects simply may not have started to present with the related effects of aging. The lower extremity aging studies generally look at individuals as old as seventy or seventy-five years old; therefore it would be most advantageous to include a subject group or groups up to seventy (70) or eighty (80) years of age.

This study could be limited with the lack of electromyography (EMG) measures of the trunk muscles. The spine joint reaction force is comprised of force contact between the intervertebral discs and vertebral bodies and all the muscles forces around the spine. An EMG recording of the trunk muscles (i.e. erector spinae) during the various task of daily living might enhance the understanding of the peak spine joint reaction forces.

A limitation in the detection of kinetic differences might be the tasks the subjects performed. Although results from the 1st through 3rd study were able to detect kinematic differences, the kinetic differences between segmented spine levels were limited. The tasks were primarily gait, thus the accelerations between segments were most likely minimal. This would inhibit the ability of the model to detect differences in joint reaction forces between spine levels. It is suggested that an activity which requires a large amount of inter-segment accelerations (e.g. sports activities) would be well suited for this model.

One additional limitation of this study was the subjects were allowed to perform the tasks at their own selected pace. This could limit the observable differences in joint reaction forces between various tasks of daily living.

Suggestions for Future Studies

Through the studies pursued in this dissertation, several aspects have been documented regarding the detailed spine motion during various tasks of daily living. These studies have also stimulated the need to pursue further work in validation of the current findings.

The first suggestion would be to use an older population of individuals. This would allow for a greater understanding of how the spine motion can change as a function of aging. Studies which study the effect of aging generally have subjects into the eighties and nineties (McGibbon & Krebs, 2001). Although this would be most interesting experimentally, this is a very frail population and the ability to test such a cohort of people may be limiting.

A second suggestion would be to control the speed at which the subjects perform the task (i.e. slow, comfortable and fast). This could yield differences in ground reaction forces, which in turn, could lead to differences in spine joint reaction forces. Such information could be used as preventative techniques in reducing back pain.

An expansion of the model to explore moment production at the various spinal segments would be most advantageous. The detailed model for the spine and trunk would show how distributions of mass throughout the trunk can alter the torque production at certain spine levels. This information could again be utilized for developing preventative techniques in reducing back pain.

The addition of a cohort of individual who are currently suffering from various forms of back pain will provide invaluable information to the how useful this procedure is at detecting motions and forces related to back pain.

Future work with direct work place application would certainly benefit from this detailed spine procedure. Individuals in the manual material handling (MMH) positions are required to handle objects that can be large in size or mass, drastically introducing an external moment into the system. This procedure could isolate what spine level would experience the most torque increase. The changes could then be added to the job in order to keep the worker safer.

Study the segmented spinal reaction forces of various athletic activities could me most advantageous. The exaggerated inter-segment spine motion which exists in many athletic endeavourers would allow for the detection of different segmented spinal reaction forces during these activities. This knowledge could be used for both training and injury prevention programs.

Back pain can be an issue with runners. To our knowledge there are not any studies that have investigated, in such detail, the motion and forces which are experienced by runners. Many training techniques could be developed from the information gathered from an investigation of runners.

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