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DEVELOPMENT, VALIDATION, AND APPLICATION OF A NONINVASIVE SPINAL MOTION MEASUREMENT SYSTEM

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ABSTRACT OF DISSERTATION

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A dissertation submitted in partial fulfillment of the
requirements for the degree of Doctor of Philosophy in
Biomedical Engineering at the
University of Kentucky.
By

Shaun Kevin Stinton

Lexington, Kentucky

Director: Dr. David Pienkowski, Professor of Biomedical Engineering

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Spontaneous vertebral fractures are a large and growing health care problem. Biomechanical factors, specifically, abnormal posture or gait-related spinal motion may interact with age-weakened bone to induce altered spinal biomechanics that in turn increase the likelihood of vertebral body fracture. This research takes steps towards the goal of reducing the number of vertebral fractures in two phases: 1) Validation of a noninvasive spinal motion measurement system in cadaver torsos and 2) Application of the measurement system in human subjects.

The cadaver study compared vertebral motion at 4 levels (T7,T12,L3,L5) as measured by adhesive skin markers versus motion measured by bone pins implanted into the vertebrae. Cadaver torsos were tested in lateral-bending, flexion and axial-rotation. Mean differences in vertebral body angular motion between skin markers and bone pin markers were <0.5° around the anterior-posterior and medial-lateral axes and <0.9° around the superior-inferior axis. This measurement method was able to accurately quantify vertebral body motion in cadaver torsos thus allowing for application to human subject testing.

X-rays and 3D motion capture were employed to quantify spinal posture and motion parameters during gait in 12 older and 12 younger normal, females. Vertebral motion around 3 axes was measured at 4 levels (T7,T10,T12,L2) using noninvasive retroreflective markers during treadmill gait at 3 speeds (0.5,0.7,0.9m/s). The average angular motion of all gait cycles at each speed was determined for each level. The tri-planar ranges of motion and variability of motion were compared as a function of age. Older subjects had 31.7% larger frontal Cobb angles and up to 30.9% and 33.5% smaller ranges of spinal motion in the frontal and sagittal planes. Variability of motion in the sagittal plane was up to 42.9% less in older subjects.

Decreased ranges of motion and variability of spinal motion observed in older subjects may imply that vertebral loading in these subjects may not be as uniformly distributed across the vertebrae as in younger subjects. Greater stresses may result from the abnormal motion, thus increasing fracture risk. Confirmation of this hypothesis

requires a longitudinal study, but if verified, may lead to the development of inexpensive countermeasures to prevent fractures.

KEYWORDS: spontaneous fractures, spinal kinematics, spinal loading, skin markers, gait-related spinal motion

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For my wife Katy, my parents, my brothers, and friends,
whose patience, understanding, and support during countless years allowed me to accomplish my goals and still be able to enjoy the time spent doing it.

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER 1: VERTEBRAL COMPRESSION FRACTURES

1.1 Vertebral Compression Fractures

1.1.1 Non-traumatic Spinal Fractures

Spinal fractures generally occur in traumatic events such as car crashes and falls, but many spinal fractures occur without a single traumatic event. These fractures are called spontaneous vertebral compression fractures and they are generally suffered by thin, middle-aged and older, Caucasian women (34,46,156). These types of fractures are characterized by vertebral body collapse and are usually seen in the thoracic and lumbar region, especially in the transition area between the two regions or thorocolumbar region (T8-L4) (32,116,154). Depending on its appearance after x-rays, the fracture can be classified as a wedge or collapse fracture (Fig 1.1).

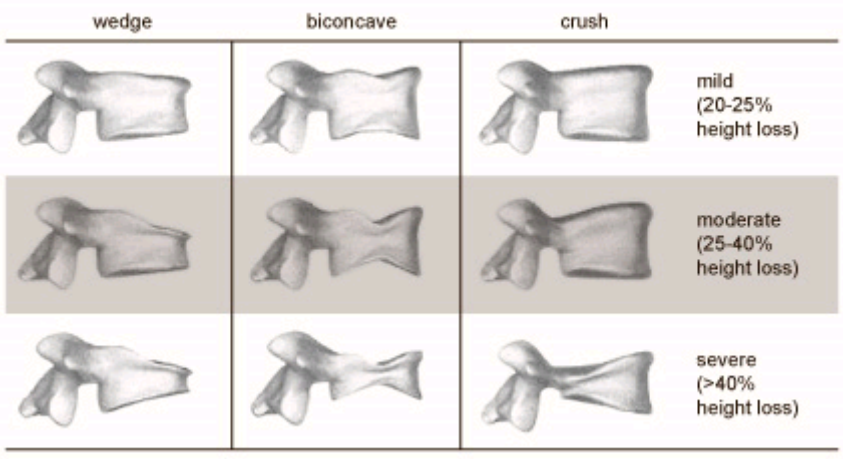


Figure 1.1. Vertebral Fracture Types

1.1.2 Etiology of Vertebral Fractures

These fractures occur in weakened bone as a result of day-to-day activities such as lifting an object, sneezing, stepping onto or off a curb, bending or twisting of the torso, or during minor falls (112,128). The bone becomes weaker with age as bone

resorption begins to overtake bone formation generally due to declining hormone levels and low calcium or vitamin D intake (28,40). The weakened bone can also result from certain drugs or diseases (40). This change in bone remodeling leads to decreased bone volume, increased porosity, decreased trabecular number and thickness, and decreased connectivity all leading to decreased mechanical strength which increases the risk of fracture (Fig 1.2) (28,40,48,104). It has also been theorized that biomechanical factors may interact with age-weakened bone to increase the risk of subsequent vertebral fracture (23).

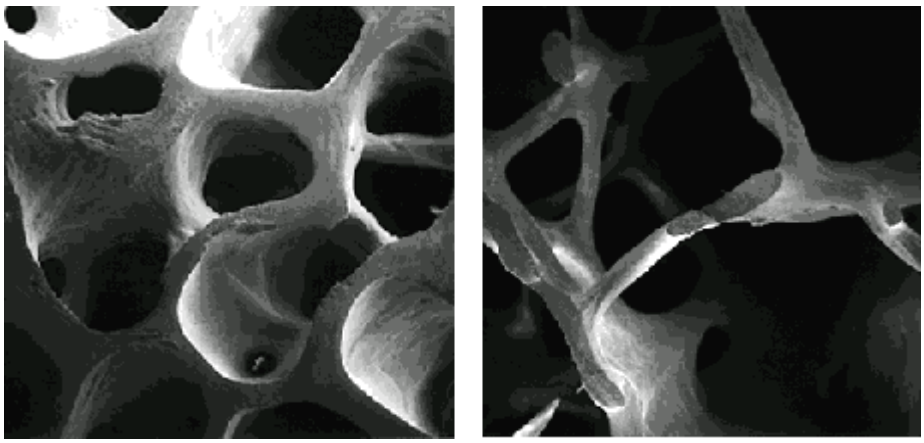


Figure 1.2. Vertebral Cross-sections in a Younger and Older Subject

Microstructure of vertebral trabecular bone in a young subject (left) and older subject (right) showing the change in bone structure.

Spontaneous vertebral compression fractures cause sudden and severe pain, loss of vertebral body height of at least 15-20%, abnormal curvature of the spine possibly leading to deformity and dowager's hump, immobility, increased number of hospital bed days, reduced pulmonary function, difficulty performing activities of daily living, and may adversely affect forward vision which may in turn contribute to falls that lead to hip or forearm fractures (1,54,89,107,120). These issues can severely alter the quality of life of subjects by causing low self-esteem, distorted body image and depression (49,94,111,127,152).

Risk factors that increase the likelihood of a fracture include: increasing age, female sex, lower BMI, family history of osteoporotic fractures, low birth weight, premature menopause, diseases such as Cushing's disease and hyperparathyroidism, smoking, drugs such as corticosteroids and heparin, rheumatoid arthritis, alcoholism, and dietary calcium or vitamin D deficiency (40,68).

1.1.3 Mechanisms of Spontaneous Vertebral Fracture

Non-traumatic vertebral fractures occur when the instantaneous load-bearing capacity of the vertebra is exceeded. The load bearing capacity of the vertebrae depends on the composition and mechanical properties of the bone, the size and shape of the bone, and the loading history.

Between 40-70% (12) of vertebral bone is trabecular bone which has altered structural capacity and reduced bone mineral density with aging as described previously. The mechanical properties and strength of the vertebral bone are important, but the time loading history of the bone must also be known to fully explain fractures. Variable loading across different portions of a vertebra can also have an effect on how the loads are transferred. The load is transferred from the intervertebral discs to the vertebral endplates and then across the trabecular structure and the thin cortical shell. The trabeculae carry the majority of compressive loads (98,138). Loading that is outside of the vertebra such as those in the soft tissues result in stresses across the trabeculae. These factors all determine the instantaneous load-bearing capacity of the vertebra. Loading that is not evenly distributed across the trabeculae can lead to varied time loading histories across the vertebrae which can lead to collapse of a portion of the vertebral body, generally seen in the anterior region.

The loads on the spine can also affected by muscular changes with aging. The stress on the vertebrae is a function of the body weight, height, and the extensor muscle force. The cross-sectional area of loading and the bone mineral density determine the resistance to these loads. The lower cross-sectional area of the vertebrae in the population that suffers these fractures (older women with low BMI) contributes

to the etiology of the fractures. Extensor muscle forces can increase due to the position of the trunk. This can be evident in posture during standing or walking. The muscle forces increase with forward body tilt. This creates a forward bending moment that must be compensated for by the muscles. Vertebral compression fractures could be due to the increased stresses caused by the increased muscle forces due to the contractions to compensate for the abnormal upper body position.

Another possible fracture mechanism in vertebral compression fractures is fatigue fracture due to repetitive loading. The modulus of the trabecular bone can decrease with fatigue as the strain accumulation increases due to creep. The vertebra can fail at lower levels of loading than would normally be required in a single instance of force application (10,55). Microdamage can accumulate in the vertebrae in a fatigue mechanism which can lead to growth of microcracks that normal bone turnover cannot repair (Wenzel).

1.1.4 Fracture Incidence

Vertebral compression fractures are the most common osteoporosis related fracture. A vertebral fracture occurs every 22 seconds worldwide, and in 2000 it was estimated that 1.4 million vertebral fractures occurred worldwide (48). Estimations of the number of non-traumatic spinal fractures have been reported for multiple populations. It has been estimated that 750,000 Americans suffer one or more non-traumatic spinal fractures each year (48,100). In Europe in 2000, there were approximately 570,000 spine fractures in persons 50 year old and older (75). In the United States, Scandinavia, Australia and western Europe, 19-25% of women over 50 have a vertebral fracture (39,64,67,100,113). The number of people suffering these fractures will be increasing as the population of women over 65 increases (100) and the prevalence of osteoporosis (40% of women by age 80 (112)) also increases. The likelihood of vertebral fracture increases with age, such that an estimated 50% of women over 80 years have a prevalent vertebral fracture (48). Estimates of vertebral

fracture incidence and prevalence taken from multiple studies (32,40,66) are shown in Table 1.1.

Table 1.1. Incidence and Prevalence of Vertebral Fractures

Age Group	Incidence Worldwide in 2000	Prevalence in Women in Saunders County, Nebraska	Prevalence in European Union Countries
50-59	158,000	11%	7%-16%
60-69	230,000	17%	12%-27%
70-79	303,000	37%	20%-43%
80+	170,000	45%	30%-76%

1.1.5 Fracture Underestimation

Vertebral fractures tend to occur earlier in life than other osteoporosis-related fractures so accurate diagnosis and treatment of these fractures is important in maintaining bone health. Despite the importance of diagnosis, only one-third of vertebral fractures come to clinical attention (25). The prevalence of spontaneous vertebral compression fractures has been shown to be greater than the reported data since many of these fractures are neither recognized nor treated (35,47). Sixty percent of women with a vertebral compression fracture do not realize they have the fracture because the accompanying pain is often dismissed as “arthritis” or “old age aches and pain” and it resolves within one to two months (108). These spinal fractures generally do not require emergency hospital care as in hip or wrist fractures which further leads to under-diagnosis. Even subjects with mild to moderate vertebral fractures who seek clinical help are often not recognized as having a fracture or the fracture is not in the medical report (25).

The proportion of vertebral fractures that go unrecognized is as high as 46% in Latin America, 45% in North America, and 29% in Europe/South Africa/Australia (35). Overall, 30-50% of vertebral fractures are misdiagnosed or not mentioned in radiology reports. This is due to either not detecting the fracture or the lack of clearly defined

language to describe vertebral fractures in the medical report (35). Only 5% of diagnosed vertebral fractures were false positives.

Another retrospective study examining lateral chest x-rays in 935 women found that 132 of the subjects (14.1 %) had a moderate to severe vertebral fracture as identified by 2 radiologists. Of the 132 fractured subjects, only 65 (52%) had a fracture reported in the radiology report. Only 23 subjects (17%) had the fracture included in the medical record or discharge summary (47).

1.1.6 Morbidity and Mortality

Vertebral fractures are associated with significantly increased mortality and morbidity, even if the fractures do not come to clinical attention. Even a single vertebral fracture can cause severe pain, loss of vertebral body height, increasing kyphosis, immobility, reduced pulmonary function, and difficulty performing activities of daily living.

Vertebral fractures are associated with an increased risk of both further vertebral and non-vertebral fractures (11,57,79,88,129,131). Women who develop a vertebral fracture are at substantial risk for additional fracture within the next 1-2 years (74,88,131). A woman 65 years of age with one vertebral fracture has a one in four chance of another fracture over 5 years, which can be reduced to one in eight by treatment (76). A vertebral fracture increases the risk of subsequent vertebral fracture by 5 times and of other osteoporotic fractures such as hip or forearm fractures by 2 times (70).

Nineteen percent of patients with an existing fracture suffer from a new vertebral fracture within one year and 25% of patients suffer another vertebral fracture within 5 years (99). A patient having 2 or more prevalent fractures has a 7 times greater risk of having another fracture within a year when compared to a non-fractured subject (48). It is estimated that up to 20% of women who suffer a vertebral fracture will suffer a new fracture within one year (48). After hospitalization for a vertebral fracture, there

is a greatly increased risk of requiring hospitalization for a further fracture in the years following initial hospitalization (65).

Increased mortality rates among women who have suffered a vertebral compression fracture have been reported by several groups. The mortality rate in the year after a fracture is 15% higher in women diagnosed with a compression fracture than those without fractures (27). In England and Wales, the reported mortality rates in women who had a vertebral fracture were 4.1% greater than the expected mortality rate after 3 months, 7.1% greater after 1 year and 13.1% greater after 5 years (157). The mortality rates increase progressively with time after a vertebral fracture up to a 16% reduction in expected five year survival rate (48). In a study that followed nearly 6500 women over an average of 3.8 years, the relative risk of death following clinical spinal fractures adjusted by age was 8.6 (18).

Subjects with multiple vertebral fractures have an increased mortality rate relative to the number of fractures with the age-standardized rates per year as follows: 0 fractures-0.02%; 1 fracture-0.023%; 2 fractures-0.025%; 3 fractures-0.028%; 4 fractures-0.032%; 5+ fractures-0.043% (70).

Vertebral fractures make daily activities such as bending forward, sitting and standing, dressing and walking up stairs more difficult. The loss of mobility due to spine fractures leads to a decrease in walking speed, may necessitate a cane or walking stick and can increase the likelihood of a fall (48). The negative impact of these issues on the quality of life is compounded by decreased independence which can lead to low self-esteem, depression and a fear of falling which can all lead to social isolation (48).

1.1.7 Fracture-related Costs

The financial cost of non-traumatic spinal fractures includes the direct costs of hospitalization, rehabilitation, and continuing care as well as indirect costs including morbidity, lessened quality of life and loss of work. In the United States, the cost of direct treatment of non-traumatic spinal fractures was estimated at $746 million in 1995 (122). This number has likely increased significantly given the advancing age of the

American population and the double-digit annual rise in healthcare costs (100). Spontaneous vertebral fractures also cause considerable loss or productivity of family members or care givers and require substantial nursing care costs. In the United States, the average hospital stay after treatment of a vertebral fracture was 6 days and more than 50% of discharged patients required some form of continuing care (100).

In 2000, it was reported that the cost of vertebral fractures in Europe was 716 million (75). The cost was projected to rise dramatically in the next 50 years. In 2007, the average cost in European Union countries for a vertebral fracture was $5,500 and the average hospital stay was 13 days (164).

1.2 Problem Statement and Purpose

1.2.1 Problem Statement

Non-traumatic vertebral compression fractures are a significant health care issue in terms of prevalence, cost and reduction in quality of life. Reduced load bearing capabilities due to osteoporosis-weakened vertebrae is insufficient to explain the etiology of spontaneous vertebral fractures. Other factors must play a role in determining who suffers a vertebral fracture and who does not.

One of these factors could be abnormal spinal loading due to abnormal motion patterns. Abnormal spinal kinematics could, over time, lead to localized weakening of the bone as a result of the altered mechanical loading (23). This abnormal loading could eventually lead to vertebral fracture when the instantaneous load bearing capability of the weakened bone is exceeded. Other possible antecedents include disc degradation and uneven vertebral endplate loading which could both be related back to abnormal spinal motion and loading (2).

1.2.2 Purpose

The ultimate goal of this research is to reduce the incidence and severity of spontaneous vertebral compression fractures. This dissertation research takes a step towards reaching this goal by examining the following specific aims:

1. To develop and validate a motion capture technique to measure spinal motion noninvasively by determining whether skin-based markers, compared to bone-pin markers, can accurately and reproducibly quantify vertebral body motion in human cadaver torsos.
2. To use x-ray data and skin-based markers to compare spinal posture and gait-related motion in older normal and younger normal female subjects to determine whether differences in parameters exist.
3. To postulate on how differences in parameters between older and younger subject groups could lead to vertebral fractures and how this could be verified and how fractures could be prevented in the future.

1.2.3 Research Plan

The rationale for the study was discussed in the first section of Chapter 1. The remainder of Chapter 1 details the previous efforts in measurement of human body kinematics in the lower extremities and in the spine.

There has not been a previous noninvasive method of spinal motion measurement using skin markers that has been validated in human subjects or cadavers. This validation is an important step to show how closely the measured motion from skin markers matches the motion of the underlying vertebrae. Therefore the first step in this research was a cadaver study using adhesive skin markers and markers attached to bone pins implanted into the vertebrae. The motion of both sets of markers was compared to determine the errors involved in using skin markers alone. This validation study is described in Chapter 2.

Once the measurement method was validated in the cadaver study, the technique was applied in human subjects. The application of the measurement technique to compare spinal posture and gait-related vertebral body motion in older normal and younger normal subject groups is described in Chapter 3.

Chapter 4 includes a discussion of the results of Chapter 2 and Chapter 3 and also the limitations of the studies, the clinical implications of the results, overall conclusions and possible future studies.

1.3 Motion Measurement

1.3.1 Lower Extremities

Skin markers with bone pins for "validation" have been used previously to provide three-dimensional (3D) motion data in the foot, ankle and the knee (4,5,7,61,62,96,97,106,124,125,162) but skin markers lack validation for quantification of vertebral motion.

1.3.1.1 Foot and Ankle

Nester et al. (106) investigated the motion of the bones of the foot using three methods: skin markers, skin markers on plates and markers attached to intracortical bone pins implanted into 9 bones of the foot. The maximum difference between any 2 out of the 3 conditions was >3° for all of the data collected and >5° in 73% of the data. This experiment shows differences between the experimental conditions, but the bone pins and skin markers or plates were not both used in the same trials. The comparison of these conditions is therefore difficult due to the differences in foot position between the trials and experimental conditions.

Westblad et al. (162) performed a similar experiment in the foot as in the Nester experiment (106). Bone pins and skin markers were used to quantify the motion of the bones of the foot, but the trials were taken separately again because the small area of the foot does not allow for both skin markers and intracortical pins to be used simultaneously. The conclusions of the study were that the superficial markers could give a reasonable measure of dorsi and plantar flexion but not eversion/inversion or add/ab duction.

In a study by Reinschmidt et al. (125), tibiocalcaneal motion was measured using markers attached to pins inserted into the tibia and calcaneus as well as from adhesive markers on the skin of the shank and shoe. The average error associated with using skin markers to measure ankle motion is presented in Table 1.1.

1.3.1.2 Knee and tibia

In a study by Stagni et al. (142), CAD models of implants were used along with a skin marker grid in knee implant subjects to compare knee kinematic data. Subjects underwent fluoroscopic evaluation during a stair climb, a step up/down, sitting to standing, standing to sitting and leg extension against gravity. The maximum artifact in the shank had a standard deviation of 21 mm and 31 mm in the thigh and rotation errors of 192% for adduction/abduction and 117% for internal/external rotation. This study showed significant artifact involved with using skin markers. The activities were limited with a confined range of motion in front of the image intensifier. There could also be errors in estimating the pose of the implants using the 2D to 3D registration.

In the study by Alexander and Andriacchi (4), an external fixation device was attached to the shank of a single subject. Markers were attached to the fixator and skin markers were also placed on the shank (cluster system-6 markers). The motion of the shank during a 10cm step up onto a platform was recorded. Using a rigid body model, the maximum translational errors ranged from 0.3 to 0.8 cm and rotational errors ranged from 4° to 8° depending on the axis of interest. An interval deformation technique was used in an attempt to minimize these errors, but the algorithm that was developed has limitations based on the changes in gait due to the external fixation device and the restriction of skin motion by the pins.

Andersen et al. (5) added kinematic constraints to the data obtained previously in a study by Benoit et al. Pins were implanted into the femur and tibia of 6 healthy subjects to determine tibio-femoral kinematics during walking trials. The overall measure error scores from using skin markers without additional constraints compared to using markers on the implanted pins ranged from $0.0228m^2$ to $0.0632m^2$ depending

on the subjects. The errors increased in all subjects when using spherical knee joint constraints and in 4 out of six subjects when using revolute knee joint constraints.

Reinschmidt et al. examined tibiofemoral motion in walking (125) and running (124). Pins were inserted into the tibia and femur and markers and markers were attached to the pins and placed adhesively to the thigh and shank. The average errors associated with using skin markers during stance phase of walking and running are presented in Table 1.1. Rotational errors up to 7.2° were observed.

Benoit et al. (7) implanted pins into the proximal tibia and distal femur and compared markers attached to the pin with skin markers. Eight subjects were recorded while performing walking and cutting motions. Average rotational differences in walking were up to 4° and average translation errors were up to 13 mm. It was determined that the skin markers could not provide representative motion of the underlying bones.

Manal et al. evaluated the accuracy of measuring tibial rotation (96) and translation (97) with skin markers during walking. Markers attached to a percutaneous device were compared to skin markers to determine the error associated with the skin markers. Rotational errors of up to 4° and translational errors up to 14.1 mm were recorded.

Houck et al. (62) implanted bone pins into the greater trochanter and tibial tubercle of 2 subjects. A femoral tracking device (noninvasive) along with tibial skin markers were used to calculate the tibiofemoral motion noninvasively. The average RMS error in using skin markers compared to bone pin markers is shown in Table 1.1.

Holden et al. (61) recorded differences in shank rotations and displacements between a percutaneous skeletal tracker implanted in the distal tibia and surface mounted markers on the mid-shank. Rotational errors of up to 8° and rotational errors of up to 10 mm were observed.

Table 1.2. Skin Marker Error in the Lower Extremities

Average angular error (degrees) of lower extremity bone motion measurements made by using bone pins versus skin markers. The ranges in the Nester study are due to the multiple bones tested.

Authors	Anatomy	Lateral Bending	Flexion	Axial Rotation
Benoit et al. (walking)	Tibiofemoral	2.5°	3.6°	2.9°
Reinschmidt et al.(walking)	Tibiofemoral	2.4°	2.1°	3.9°
Reinschmidt et al.(running)	Tibiofemoral	4.1°	5.3°	4.4°
Reinschmidt et al. (walking)	Tibiocalcaneal	2.5°	3.1°	3.4°
Houck et al. (walking)	Tibiocalcaneal	1.5°	1.3°	1.0°
Andersen et al. (walking)	Tibiofemoral	2.1°	1.7°	2.7°
Manal et al. (walking)	Tibia	1-2°	1-2°	3-7°
Nester et al. (walking)	Multiple bones in the Foot	2.3-3.9°	2.6-3.7°	1.9-5.1°

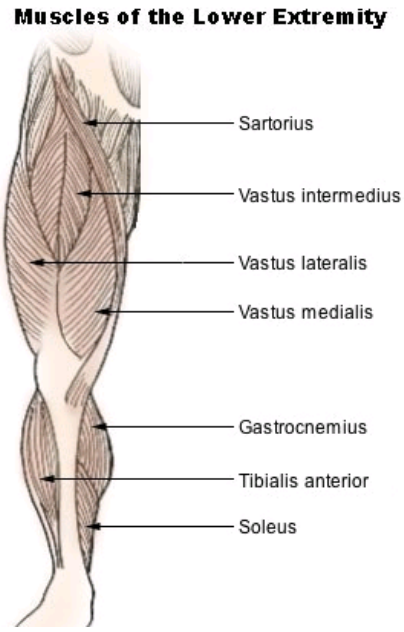
1.3.2 Back Anatomy vs. lower extremities

There are several issues related to ankle anatomy that make noninvasive tracking of motion very difficult. There are not easily definable bony prominences that markers can be placed over in the talus and navicular. The large variation in anatomy of the foot between subjects and the small size of the bones involved are also difficult problems to overcome (Figure 1.3) (93).

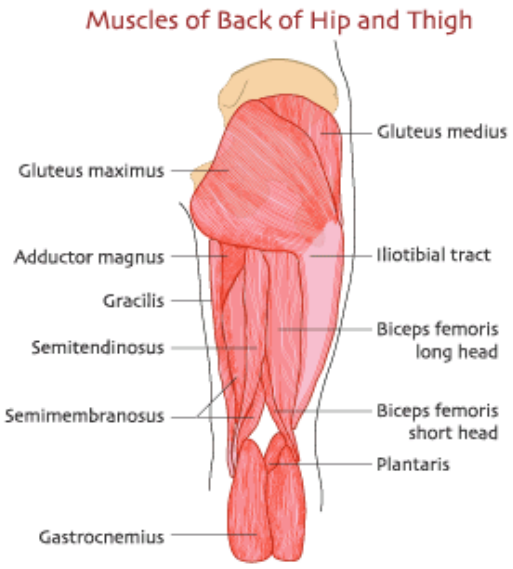
The motion of the tibia can be measured well using skin markers, however, the motion of the distal femur cannot be accurately measured with skin markers, especially in rotation. This is due to the large amount of muscle in the quadriceps and hamstring that is between the skin and bone (Figure 1.3). The skin covering the patella is also loosely attached to the deeper soft tissues allowing for a large amount of skin sliding over the bone (93). The deep location of the proximal femur makes skin marker analysis of hip motion more difficult as well.

The anatomy of the back shown in Figure 1.4 was thought to allow for skin marker analysis due to the manner that the skin is attached to the deeper soft tissues and the fact that there is not a large mass of muscle in the area of the spinous processes. The back fascia is rigidly attached to the spinous processes so the skin will follow movement of the vertebrae better than in the bones of the lower extremities (93).

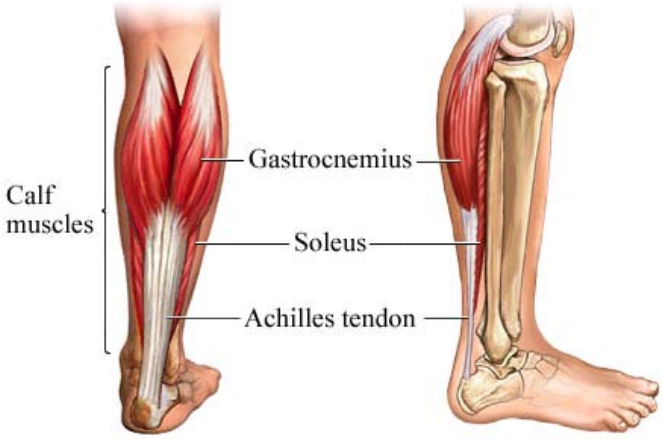
There is not a large mass of muscle or fat around the spine and there are many ligaments and fascia along the spine and between spinal levels connecting the skin to deeper structures that should allow for better results (less skin motion) than using skin markers in other areas of the body. The average differences found between bone pin angles and skin marker angles in other areas of the body are shown (Table 1.1).



Leg muscles-anterior



Upper leg-posterior



Lower leg-posterior

Figure 1.3. Anatomy of the Lower Extremities

Anterior and posterior views showing the large muscles over the bones of the thigh and shank.

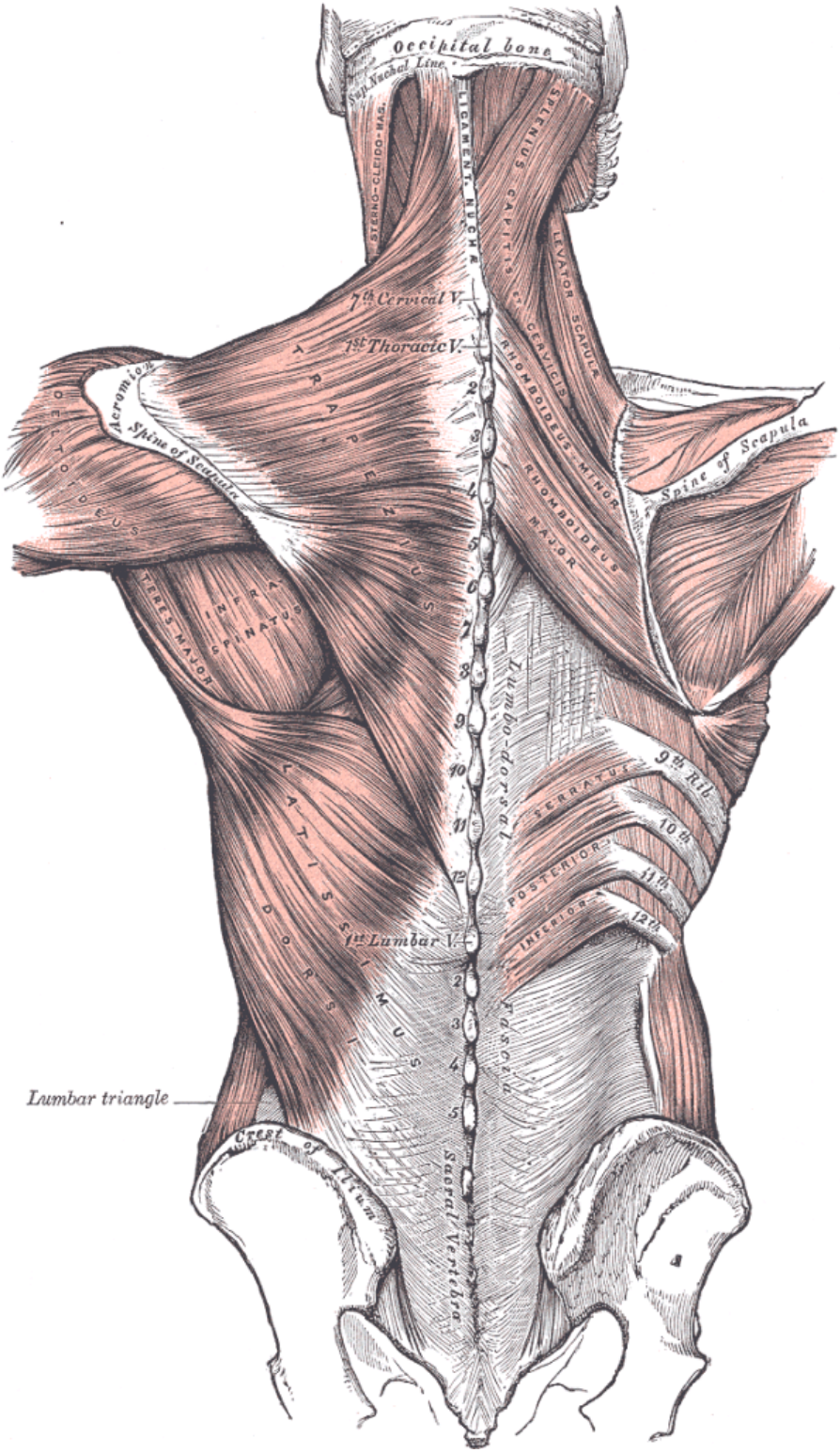


Figure 1.4. Anatomy of the Back

1.3.3 Invasive Measurement of Spinal Motion

Invasive bone pins have been used to quantify spinal motion (51,91,92,133,134,144). These techniques remain the gold standard for vertebral motion measurement because such pins are implanted directly into the vertebral body. Invasive bone pins, however, are unfeasible for a large scale clinical study due to technical difficulties, infection concerns, and subject recruitment issues.

Gregersen and Lucas (51) measured axial rotation in the lumbar spine while standing, sitting and during normal walking in seven male subjects. Bone pins were inserted into varying levels of the thoracolumnar spine and motion was measured using relative-rotation transducers. There were no levels with more than 3 subjects. A limitation of the study was that the pin insertion was not consistent due to comfort issues with the subjects. The axial rotation of the individual spinal levels ranged from 2°-9.4°, but these extremes were at levels that only had a single subject with a pin implanted there.

Lumsden and Morris (91) investigated lumbosacral motion using bone pins implanted into the spinous process of L5 and into both posterior superior iliac crests of nine subjects. Relative-rotation transducers were used to measure axial motion during walking and the rotations observed were 1.1°-1.93°.

Lund et al. (92) implanted 2 percutaneous transpedicular screws into the affected levels of each of 34 subjects, usually at L4 and S1 or L5 and S1. Markers were attached to each screw and also one marker was adhesively placed on the skin and the markers were tracked by a camera system. The positions of the markers were recorded during the active motions of flexion/extension, lateral bending and axial rotation. The results from the study were compared to data from the literature for normal subjects. The range of motion in the low back pain subjects was smaller than the normal subjects as expected due to the pain threshold. Differences were found in the flexion/extension ratio, the lateral bending asymmetry and the coupled axial rotation-lateral bending ratio. This study can accurately record the motion of the vertebrae, but it only examines subjects with low back pain and has no results showing the motion in normal spines.

Rozumalski et al. (133) implanted bone pins into each level of the lumbar spine (L1-S1) in 10 healthy subjects. Marker triads were attached to each pin and the positions of these markers were tracked during walking trials and also in the active motions of flexion/extension, lateral bending and axial rotation. The motion was reported intersegmentally from a superior vertebra to the inferior vertebra with the inferior vertebra being fixed. The average motion for all levels in walking ranged from 4.26°–4.38° in the sagittal plane, 2.61°–4.00° in the coronal plane, and 4.11°–5.24° in the transverse plane. This study is beneficial as it examines healthy subjects during walking trials. The subjects were also limited to those with a BMI<25 which lines up well with the population that suffers from vertebral compression fractures.

Sahni et al. (134) implanted transpedicular pins into 2 levels (L3 and L4) of 6 cadaver spines. Markers were attached to the pins and other markers were also attached directly to the vertebrae. The cadaver spines were placed in flexion, extension and left and right lateral bending and the marker sets were compared. There was an excellent correlation between the marker sets (r^2=0.931) showing that markers attached to bone pins are indeed an accurate representation of the motion of the vertebrae. This study is important as it shows the accuracy of using markers attached to bone pins implanted into the vertebrae. This is important because in the cadaver study the markers attached to bone pins were used as the gold standard to which the adhesively attached skin markers were compared.

Steffen et al. (144), inserted pins into the spinous processes of L3 and L4 in 16 healthy men. Electromagnetic tracking devices were attached to the pins and the motion of the L3 vertebra relative to the L4 vertebra was measured during active flexion/extension, lateral bending and axial rotation. Consistent coupled axial rotation motion was found during lateral bending with left axial rotation occurring during right lateral bending and vice-versa. Mean ranges of motion for flexion/extension, lateral bending (one side) and axial rotation (one side) were: 17°, 6.3° and 1.1° respectively. This study helps further the knowledge of how the vertebrae move in normal subjects,

but the subjects in this case were all male. It also reports data during active motions but not during gait.

These invasive methods provide the most accurate measure of the true vertebral motion but were ruled out in this research due to the difficulty in passing the research through the IRB, the difficulty in finding female younger or older subjects willing to have bone pins implanted into their spines, the cost and risk involved in the implantation of bone pins, and the increased payments to subjects. Various noninvasive methods were then examined to determine the best fit for this research.

1.3.4 Noninvasive Measurement of Spinal Motion

Several different methods of measuring spinal motion noninvasively have been used previously including imaging (43,56,63,72,87,102,109,115,117,118,119,150,167), electromagnetic systems (32,85,132,166,168) and adhesive markers (3,19,20,21,22,29,30,41,45,50,71,82,86,137,147,151,158,159,161,163,169,170,171).

1.3.4.1 Imaging Techniques

Frobin et al. (43) measured segmental motion in the lumbar spine from T12 through S1 using radiographs. The technique used digitized landmarks in an attempt to increase the accuracy of the method. The errors of the method were found to be 0.7°-1.6° for angle measurement and displacement error was 1.2%-2.4% of vertebral body depth with the ranges due to the multiple levels that were tested. Only sagittal plane angles were determined in this study.

Harada et al. (56) measured active flexion/extension from the L3-L4 level through L5-S1 in the sagittal plane with 10 subjects using cineradiography. In forward flexion, the motion started at the L3-L4 level and continued to the lower levels with phase lags and with angular velocity increasing at lower levels at the onset of flexion. In backward flexion, the motion started at the L5-S1 level and there was not a relationship between level and angular velocity. The range of motion was approximately 16° from each vertebra relative to the inferior vertebra in both forward and backward flexion.

The imaging was done continuously so it was better than single still images, but the range was still limited by the size of the image intensifier which may not allow for normal spinal flexion.

Ishii et al. (63) examined the upper cervical spine using MRI taken at 15° increments during head rotation. Mean maximum angles in axial rotation were 1.7° and 36.2° to each side between the Occipital level-C1 and C1-C2, respectively. Coupled lateral bending and extension occurred with axial rotation with the lateral bending being in the opposite direction of the axial rotation. The upper cervical spine was not the region of interest in this dissertation research since it is not a location of vertebral compression fractures. The data collection was also done at finite points which is not necessarily the motion seen in normal activities.

Kanayama et al. (72) examined intersegmental motion from L3-L4 to L5-S1 in 8 subjects using similar method as Harada et al. Motion lags were observed during flexion with the motion starting at the L3-L4 motion segment. The L4-L5 was an average of 6° delayed from L3-L4 motion. The L5-S1 motion was 8° delayed on average from the L4-L5 motion. The total range of motion between L3 and S1 averaged 37°.

Lim et al. CT scanned cadaver cervical spines (C2-C3) in 5° and 20° of axial rotation, lateral bending and flexion to measure the relation of the vertebrae to each other (87). The noninvasive system was accurate to within 1° in rotation and 1mm in translation.

Mörl et al. (102) used open MRI of the L3 and L4 levels to examine skin marker position related to the spinous processes in 5 seating positions. The relation between marker location and underlying vertebrae had Pearson's correlation coefficients between .916 and .993. The relation between the translation vectors in the external markers and in the MRI measurements had Pearson's correlation coefficients between .68 and .99. The angular orientations of the vertebrae had correlations between .769 and .959 between the MRI and skin markers. The measurements in this study were taken in static positions which is limited for comparison to the dynamic nature of gait.

The subjects were also young, fit males whose soft tissue properties would not be the same as the older females who typically suffer spontaneous vertebral fractures.

Three-dimensional lumbar spine segmental motion from L1 through S1 was measured in 15 normal subjects by Ochia et al. (109). The subjects were imaged with a CT scanner while lying supine and while axially rotated 50° to the left and right. More axial rotation motion was observed at the upper levels (L1 and L2). More axial rotation was observed in subjects in their 40's compared to subjects in their 30's. Complex coupling motions were also observed in axial rotation. The magnitude of the lateral bending motion was the largest and lateral bending was observed in the opposite direction of axial rotation.

Panjabi et al. (115) loaded nine cadaveric lumbar spines with pure moments of flexion/extension, axial rotation and lateral bending and the intervertebral motions were determined at each level (L1-L2 through L5-S1 by using stereophotogrammetry. The flexion motion ranged from 8-15° with larger motion at the lower levels. The axial rotation motion ranged from 1-2° and the lateral bending motion ranged from 4-6°.

Pearcy et al. (117,118) measured the normal motion of the lumbar spine in flexion, axial rotation and lateral bending using X-ray analysis in 11 subjects. The L1-L2 through L5-S1 levels were imaged. The average total range of motion of each intervertebral level in flexion and extension combined was 14°. There was an average of 2° of axial rotation at each intervertebral level with the L3-L4 and L4-L5 levels having slightly larger ranges. In lateral bending, the upper 3 levels averaged 10° of motion while L4-L5 and L5-S1 had less motion: 6° and 3° respectively. Axial rotation was found to be coupled with lateral bending to the opposite direction in the upper 3 levels but in the L5-S1 level the bending was in the same direction as the rotation.

A study Plamondon et al. (119) was similar to the studies by Pearcy et. al. The same motions were imaged with stereoradiography in the lumbar spine of 16 normal subjects. The mean relative displacements of the lumbar vertebrae were measured to be 10° in flexion, 3° in extension, 5° in lateral bending and 1° in axial rotation.

Teyhen et al. (150) used fluoroscopy to record flexion motion in the lumbar spine of 20 subjects: 11 with back pain and 9 without back pain. An algorithm was applied to determine segmental angular and linear displacement. Interimage and intraimage reliability were determined using intraclass correlation coefficients. Interimage correlations averaged 0.878 with standard error of 0.4-0.7° and 0.2-0.3mm while intraimage correlations averaged 0.986 with standard error of 0.7-1.4° and 0.4-0.7 mm. This study only examines flexion. The techniques may not be applicable in other motions or in coupled motions.

The study by Wong et al. (167) was similar to the study by Teyhen et al. in that flexion/extension motion of the lumbar spine was measured using video fluoroscopy. This motion was recorded from 20° of flexion through 10° of extension in 30 healthy subjects from lower endplate of the L1 vertebra to upper endplate of the S1 vertebra. The range of motion was greatest at the L1-L2 level and decreased at each inferior level.

Some of these studies offer good accuracy in determining spinal kinematics but imaging may not be the best method for a large study due to the exposure to radiation, the limited range available for the full range of motion, poor or inconsistent image quality and the cost associated with the testing.

1.3.4.2 Electromagnetic Techniques

Feipel et al. (41) measured kinematics of the lumbar spine in 22 normal subjects during treadmill walking at 4 speeds (0.8, 1.1, 1.4 and 1.7 m/s). An electrogoniometer was used to measure the motion between T12 and the sacrum. No significant differences were found between male and female or in age differences. Average sagittal plane range of motion was 6% of the maximum active flexion/extension range. Frontal plane range of motion during walking was 13-18% of the active lateral bending range. Average transverse plane range of motion was 21-37% of the active axial rotation range. Transverse plane range of motion increased with walking speed. Coupling of rotation and bending was individually variable and dependent on walking speed.

In a study by Lee and Wong (85), lumbar spine and hip motion was measured in 20 healthy subjects using electromagnetic sensors attached to the spinous processes of L1 and the sacrum and to the left and right posterior thigh. The subjects performed flexion and extension, lateral bending and axial rotation. The spine and hips both contribute in flexion and extension motions, the spine is the primary source of motion in lateral bending and the hips are the predominate source of motion in twisting.

Rowe and White (132) used an Isotrak magnetic tracking system to measure lumbar spine kinematics during walking of 10 nurses who had experienced mild back pain. The group average curves ranged from 3.7° to -2.9° in axial rotation, 1.8° to -2.2° in lateral bending and 6.1° to 3.8° in flexion. Average curves were reported showing the gait patterns for flexion/extension, lateral bending and axial rotation. These patterns are useful for comparison with the patterns observed in this dissertation research.

In an extension of the earlier study, Wong and Lee (166) placed 2 electromagnetic sensors over the L1 level and the sacrum in 20 normal subjects, 24 back pain subjects with no limitations in leg lift and 17 back pain subjects with limitations during leg lift. Two other sensors were placed on the posterior left and right thigh to measure hip motion. The subjects performed forward and back bending, side bending and axial twisting and motion of the lumbar spine and hip was recorded using the electromagnetic sensors. The mean lumbar spine range of motion in the normal subjects was 61.9° in flexion, 15.5° in backward bending, 23.7° in lateral bending and 12.2° in axial rotation. Back pain limited the range of motion in all directions in the spine and in hip flexion.

Yang et al. (168) measured lower thoracic and lumbar spinal motion with an electromagnetic system and validated the system with radiography. The study was looking at osteoporotic spines so the same population was investigated as in this dissertation research. Sensors were placed over the spinous processes of the T7, L1 and S1 vertebral levels. Radiographs were taken at neutral, full flexion and full extension with the electromagnetic sensors attached to validate the electromagnetic motion values. The accuracy of the electromagnetic system by itself was found to be poor. The

differences in rotation angle between T7 and L1 were 5.62% from flexion to neutral and 12.3% from extension to neutral. The differences in angle between L1 and S1 were 8.31% from flexion to neutral and 10.85% from extension to neutral. This method was only used in flexion and extension motions so it is limited in what can be applied to walking motions.

1.3.4.3 Skin Marker Techniques

Al-Eisa et al. (3) collected kinematic data in 59 normal subjects and 54 subjects with a history of low back pain. Thirteen markers were used to define 4 segments: cervical spine, thoracic spine, lumbar spine and sacrum. The subjects performed lateral bending and axial rotation motions. No difference in overall range of motion between the groups was observed, but more asymmetry was evident in the low back pain subjects.

Cerveri et al. (19), placed adhesive markers over the spinous and transverse processes of the T11 vertebra through the L5 vertebrae and on the pelvis. Motion was recorded during flexion, lateral bending and axial rotation. A kinematic model was developed in an attempt to account for the skin sliding. All skin sliding correction was done based on simulation and no laboratory validation testing was done with bone pins or imaging.

Chockalingham et al. (21,22) examined optimal marker placements in the back and measured relative motion between the back and lower limbs. Single retroreflective markers were place from C7 through S3 at every other level and motion capture cameras recorded the markers during flexion and lateral bending. This study was a pilot study and did not include walking.

Gal et al. (45) measured vertebral translations during manipulative therapy using bone pins, skin markers and accelerometers. The experiment was done at the T10, T11 and T12 levels of two cadavers. Posterior to anterior translations were recorded with all three methods and compared. There were no significant differences in absolute or relative translations between the bone pins and the skin markers. The accelerometers

overestimated the absolute translations and underestimated the relative translations. This adds to the evidence that skin markers can be effectively used in the thoracic and lumbar spine. However, this experiment was completed in cadavers, which do not represent the true behavior in living persons.

Gracovetsky et al. (50) used light emitting diode markers on the spinous processes and iliac crests in 40 subjects to measure the motion of the spine during flexion and lateral bending while unloaded and with a series of weights. Bilateral electromyography of the multifidus was collected simultaneously at the L5 level. Motion patterns were consistent and varied little with load while older subjects did have less mobility. This study compared values to previous radiographic studies, but no validation experiments were carried out. The data obtained from skin markers did match up well with previous radiographic studies though.

Kyphosis and lordosis were measured noninvasively using methods developed by Leroux et al. (86). Single reflective markers were placed on the spine over the spinous processes of T1, T3, T5, T7, T9, T11, L1, L3, L4, L5 and S1. The mean absolute differences between the angles measured by the skin markers and the radiographic measurements were 5° in kyphosis and 6° in lordosis. This technique measured angles from a lateral radiograph so analysis was only two dimensional. Single markers on each spinal level would not be as effective in the three dimensional motion of the spine during walking.

Schache et al. (137) measured lumbar spine and pelvis angular kinematics in 20 male subjects during running at 4 m/s. Motion patterns of the pelvis and lumbar spine were reported graphically and were shown to have a complex relationship during running.

Zhang and Xiong (169) performed model guided derivation of lumbar spine kinematics using a model along with skin markers. Markers were placed at C7, T7, L2, L3, L4, L5, and S1 and subjects performed a lifting task. The model attempted to find the centers of rotation of the vertebrae.

Chan et al. (20) attempted to assess the repeatability of measuring trunk motion using external markers and the repeatability of spinal motion during walking in normal

and scoliotic adolescents. Motion of the shoulder, spine and pelvis was recorded. Shoulder motion in all three planes was not very repeatable. Repeatability of spinal motion in the frontal plane was fair, but the repeatability in the sagittal plane was poor. Pelvic coronal and transverse plane motion was very repeatable, but the sagittal plane was not. It was also found that the scoliotic adolescent patients had less variable spinal motion. This method used single markers on the spinal levels so spinal rotation would be difficult to measure accurately.

Crosbie et al. (29,30) examined lower thoracic and lumbar spine spinal kinematics in 108 healthy adults. Thirteen markers were placed adhesively on the back along with markers on the thighs and feet. Subjects walked at 2 speeds: a comfortable speed and the speed they would walk if they were in a hurry. An increased range of motion was observed with increased walking speed. A significant reduction in spinal range of motion was observed with increasing age. Peak to peak ranges of motion in lateral bending, flexion and axial rotation were: Lower thoracic (7°, 2.5°, 4°), lumbar (9°, 3.5°, 4.5°) and pelvis (6°, 3.5°, 4°). Typical patterns of spinal motion were graphically represented which can be compared to the data obtained in this dissertation research.

The effect of leg length discrepancy on spinal kinematics was investigated by Kakushima et al. (71). Twenty two healthy, male subjects were fitted with a heel raising device for the right foot. Single reflective marker was placed over the spinous process of every other spinal level from T1 through L5. The motion of the spinal levels was recorded during walking with and without the heel raising device. The average maximum lateral bending to one side ranged from 2.3° to 6.2° with the larger ranges found in the lumbar spine. The ranges increased in trials that used the heel raising device.

Konz et al. (82) developed a kinematic model to assess spinal motion during gait in 10 normal subjects. Single reflective markers were placed over the spinous processes of C1, C7, T3, L1 and S1 while marker triads were used at C5, T7 and L3. The average flexion ranges of motion of the 10 normal subjects during gait were 7.3°, 5.7° and 4.1° for the cervical, thoracic and lumbar spine. The lateral bending ranges of motion were

4.0°, 8.7°and 5.7°, while the axial rotation ranges were 12.5°, 10.8° and 9.8°. The marker setup for this experiment is close to what was ultimately used in this dissertation research, but did not use marker triads at each level of interest. The model was validated to a mechanical model to verify the accuracy of the camera system, but there was no cadaver or imaging model to show that the system was measuring the accuracy in measuring vertebral motion rather than skin motion.

Stokes et al. (147) measured spinal kinematics in 8 subjects during treadmill walking at various speeds. The motion patterns were complex with similar patterns over subjects. The total range of motion was found to increase with walking speed.

Thurston and Harris (151) recorded the motion in the lumbar spine and pelvis during walking in 48 male volunteers. Markers were attached to rigs over the upper lumbar spine and sacrum. The rigs used were large and there was a band wrapped around the front of the subject to secure the rig which could alter the kinematics recorded. The average motion of the pelvis during walking was 4.1° in the sagittal plane, 8.5° in the coronal plane and 8.3° in the transverse plane. The average motion in the spine during walking was 4.1° in the sagittal plane, 7° in the coronal plane and 10.1° in the transverse plane. The duration of the stance phase was observed to be longer in older subjects when compared to younger subjects.

Vogt investigated lumbar spine and pelvis kinematics during walking in several studies (158,159,161). The T12 level and the pelvis were tracked using ultrasonic markers on the skin at the select levels. Correlation values ranged from .76-.98 showing the close relationship between motions of the pelvis and lumbar spine. In one study comparing low back pain subjects to normal subjects, higher variability in stride-to-stride movement was found in the abnormal subjects. Another study found statistically significant differences in sagittal plane movement were shown between over-ground versus treadmill walking.

Whittle and Levine (163) recorded lumbar spine and pelvis kinematics in 20 adult males during normal gait. Rigs with reflective markers were attached to the thoracolumbar junction and the upper sacrum. The total range of motion of the lumbar

spine was 3.8° in flexion, 7.55° in lateral bending and 8.34° in axial rotation. The sagittal plane showed the most variability between subjects. The average curves for flexion, lateral bending and axial rotation were also presented graphically which is useful for comparison to the kinematic data from the current dissertation research.

Zhao et al. (171) studied the kinematics of the entire spine during locomotion. Reflective markers were placed on the skin at C1, C4, C7, T3, T5, T7, T9, T11, L1, L3 and S1. Segmental coupling of the spine in the transverse and frontal planes was observed during walking.

Attempts to quantify vertebral motion noninvasively have used imaging techniques, electromagnetic techniques, and skin marker techniques. None of these previous methods was deemed to be sufficiently accurate or adequately validated for the quantification of vertebral body motion during walking. Therefore, a new technique was needed which included the necessary validation.

After examination of the available techniques for noninvasive quantification of spinal motion, a skin-based marker system that had been validated in a cadaver model would be the best choice for this research. The subjects will be able to move in a more natural manner using adhesive skin markers rather than using imaging with a confined space or electromagnetic systems where the necessary wires can alter the normal gait patterns. The adhesive skin markers are also cost-effective and allow for examination of a large number of subjects in a short time period for each subject. A system of adhesive markers and cameras would not be as effective without the cadaver validation though because the knowledge of how closely the skin motion mirrors the motion of the underlying vertebrae is vital in the validity of the results.

CHAPTER 2: DEVELOPMENT AND VALIDATION OF A NONINVASIVE SPINAL MOTION MEASUREMENT SYSTEM

2.1 Introduction

It has been estimated that 750,000 Americans, including one in four post-menopausal women, will suffer one or more low-energy vertebral fractures each year due to osteoporosis (100). These fractures can lead to loss of height, abnormal curvature of the spine, pain, disability, and disruption of vision due to increased kyphosis thereby predisposing to falls and fractures of the appendicular skeleton. One-fourth of those who suffer one vertebral fracture will suffer another vertebral fracture within 5 years (99). The mortality rate in the year after a fracture is 15% higher in women diagnosed with a vertebral fracture than in a control population (27). Low-energy vertebral fractures were associated with direct healthcare costs of $746 million in 1995 (100). The prevalence of vertebral fractures will increase with the growing population of women over 65 years of age and the increasing incidence of osteoporosis (40% of women by age 80 years old) (112).

Vertebral compression fractures are commonly believed to occur due to a single large stress, such as that resulting from a fall or motor vehicle crash. In the elderly, however, such fractures can occur with minimal force application, i.e. those accompanying lifting a light household object, alighting from an automobile, sneezing or coughing (112,128). These fractures are sometimes referred to as spontaneous fractures because the loads applied are those incurred during the activities of daily living. They are generally wedge-shaped and occur at the anterior portion of the vertebral body, principally in the thoracolumbar region between the tenth thoracic (T10) and second lumbar (L2) vertebrae (154) and are most common in thin, middle-aged and older, Caucasian women (34,46,156).

Loss of bone mass through osteoporosis is one factor believed to have a role in the etiology of low-energy vertebral fractures; however, these fractures cannot be

explained by low bone mass alone. Other possible antecedents include disc degradation and uneven vertebral endplate loading (2). The central hypothesis of the present research is that there is an additional mechanical factor, specifically, abnormal spine loading accompanying abnormal posture and dynamic (gait) motion, which when applied cyclically to mechanically inferior (osteoporotic) bone could result in a cascade of structural degradation that culminates in the observed vertebral compression fracture. To reduce the incidence and severity of spontaneous vertebral compression fractures, it is first necessary to evaluate this hypothesis and determine if specific postural patterns or gait-related spinal motions are associated with increased vertebral loading. If these patterns can be identified, then countermeasures such as orthotics, targeted strength or balance training could be employed to prevent or delay the onset of non-traumatic vertebral fractures.

Invasive bone pins have been used to quantify spinal motion (92,133,134,144). This technique remains the gold standard because such pins are implanted directly into the vertebral body. Invasive bone pins, however, are unfeasible for a large scale clinical study due to technical difficulties, infection concerns, and subject recruitment issues.

Non-invasive methods have been developed to measure vertebral motion using imaging techniques (56,63,72,102,115,117,118,150,167,171) or skin markers (3,19,21,22,29,30,41,45,50,71,82,86,110,147,158,159,161,163). Skin markers with bone pins for “validation” have been used previously to provide three-dimensional (3D) motion data in the foot, ankle, and the knee (5,7,62,106,124,125), but skin markers lack validation for quantification of vertebral motion. Thus, the objective of the present study was to determine whether skin-based markers, compared to bone-pin markers, can accurately and reproducibly quantify vertebral body motion in human cadaver torsos.

2.2 Material and Methods

2.2.1 Study Design

A laboratory cadaveric study was performed to quantify the error in using adhesive skin markers to measure vertebral motion as compared to using bone pins implanted into the vertebrae. Five trials of each of five motions were tested: flexion/extension, left lateral bend, right lateral bend, left axial rotation, and right axial rotation. Each trial was its own control because both adhesive markers and bone pins were used to calculate the angles during each trial.

2.2.2 Cadaver Specimens

The donor cadaver torso inclusion criteria were: either sex, ≥55 years old, low-normal to underweight body mass index and negative for HIV and Hepatitis. Three cadaver torsos were received non-embalmed, partially thawedand eviscerated (Table 2.1). The torsos were received in the afternoon and were left in a refrigerator overnight to finish thawing completely. The testing was completed the morning after each torso was received. The testing lasted approximately 3 hours including setup. Fresh frozen cadaver torsos were used in order to have the skin properties as close to in vivo conditions as possible.

Table 2.1. Cadaver Torso Characteristics

Sex	Age	Height (m)	Weight (kg)
Female	95	1.5	41.7
Male	71	1.7	45.4
Female	59	1.6	40.8

2.2.3 Markers

Stainless steel bone pins of 0.125m length (0.004m diameter Schanz Screws, Synthes Inc., West Chester, PA) were inserted by a fellowship trained Orthopaedic Spine surgeon through the skin of the torso into the pedicle of each of four vertebrae: seventh thoracic vertebra (T7), twelfth thoracic vertebra (T12), third lumbar vertebra (L3) and

fifth lumbar vertebra (L5). A second bone pin was inserted into the pedicle of the T7 level to verify that the two pins in the same rigid body provide equivalent motion. The skin and underlying tissue were cut at the base of the pin to minimize restriction of skin motion. Skin marker triads (3 retroreflective markers on 3 pins all secured into a single base) were attached to each pin. The retroreflective marker triads consisted of three spheres (0.005m diameter) placed 0.04m apart in an equilateral triangle. A fourth marker was placed in the center of the triad on top of the bone pin (Figure 2.1). Pilot testing established that 0.04m marker separation was an optimal compromise between compactness (required by the size of the vertebrae) and sufficient distance to minimize errors in the calculated angles due to motion analysis camera position.

Skin marker triads (identical to the bone pin triads) were placed adhesively on the skin over the spinous processes of the same levels (T7, T12, L3, L5). Eight markers were placed on the skin over the left and right transverse processes and four markers were place on the skin over the pelvis (both iliac crests and posterior superior iliac spines). Three additional markers were placed on the back (one on upper right and two on the left) to assist in marker identification. The transverse process markers were used to aid in definition of local coordinate systems for each spinal level. A total of 51 markers were placed on each torso (Figure 2.1).

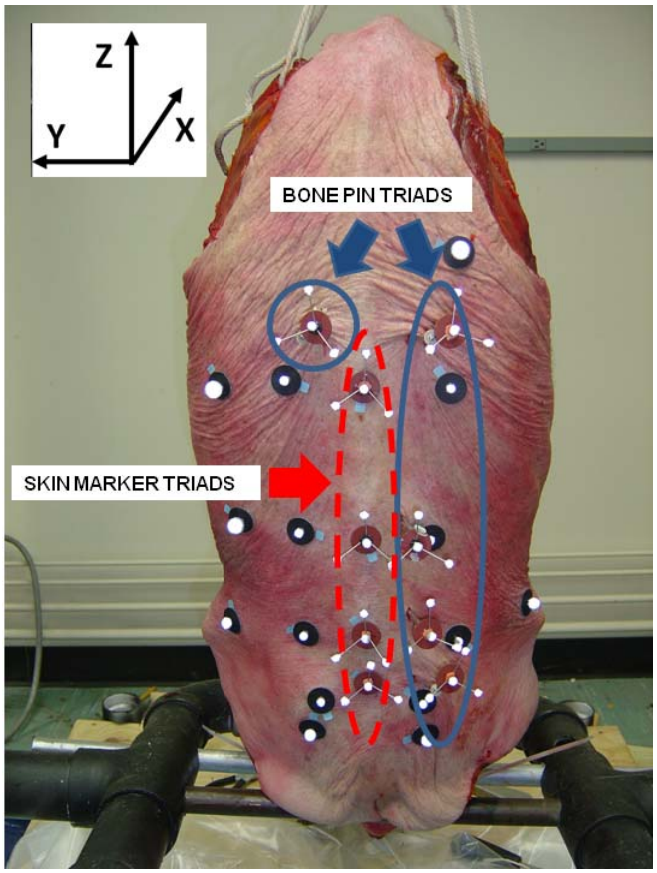


Figure 2.1. Cadaver Torso Mounted to the Fixture

A cadaver torso attached to the testing fixture. The skin marker triads (dashed) and bone pin triads (solid) are indicated, and the remaining transverse process markers, pelvis markers, and offset markers are also visible.

2.2.4 Testing Apparatus

A testing apparatus was custom-built to hold each torso and allow for physiologically relevant lateral-bending (x), flexion/extension (y) and axial-rotation motions (z); (Figure 2.2). Torsos were attached to the apparatus at the pelvis to minimize pelvis motion relative to the fixed laboratory coordinate system. The ischium rested on top of and between two metal bars on the testing apparatus. Stainless steel screw clamps were placed through each obturator foramen and were used to fasten each pelvis to the metal bars. A wooden block was attached to the sternum with bone screws and to achieve flexion/extension motion, a bar was attached to this sternal block

and then to a linear slider on the base of the apparatus which moved in the anterior-posterior direction. A stop in the slider was used to control the maximum flexion and extension. Axial-rotation was achieved by connecting a line to the sternal block and then this line was routed through a pulley attached to the apparatus at the same height and approximately 0.40m lateral to the torso. Lateral-bending was achieved by using a medial-lateral slider approximately 0.40m above the torso. The range of the motion was controlled with a stop in the lateral slider that allowed for consistent trials.

2.2.5 Data Capture

Seven Eagle-4 cameras (2352x1728 pixels; Motion Analysis Corporation, Santa Rosa, CA) were placed in a semicircle approximately 2.5m behind the torso to view a volume of 0.5m x 0.5m x 1m (x, y, z) in which the markers would be moving. Motion of the retroreflective markers within this volume was calibrated by using a seed and wand method and data capture software (Cortex 1.0, Motion Analysis Corporation). The seed was an L-frame with 4 retroreflective markers located at precisely spaced distances on this frame to define the laboratory coordinate axes. The wand was a rod with 3 retroreflective markers mounted at known distances and this wand was waved through the specified volume. Mean (± SD) linear spatial calibration errors were 0.08 ± 0.036 mm. Two-dimensional data from each camera was captured at 60 Hz for five motions: left/right lateral-bending (approximately 15°), flexion (approximately 20°) and left/right axial-rotation (approximately 11°). The two-dimensional data were used along with the calibration data and the focal lengths of the camera to determine the three-dimensional coordinates of each marker throughout each motion. Each of the 3 torsos studied were subjected to 5 trials of each of these 5 torso motions (75 total tests).

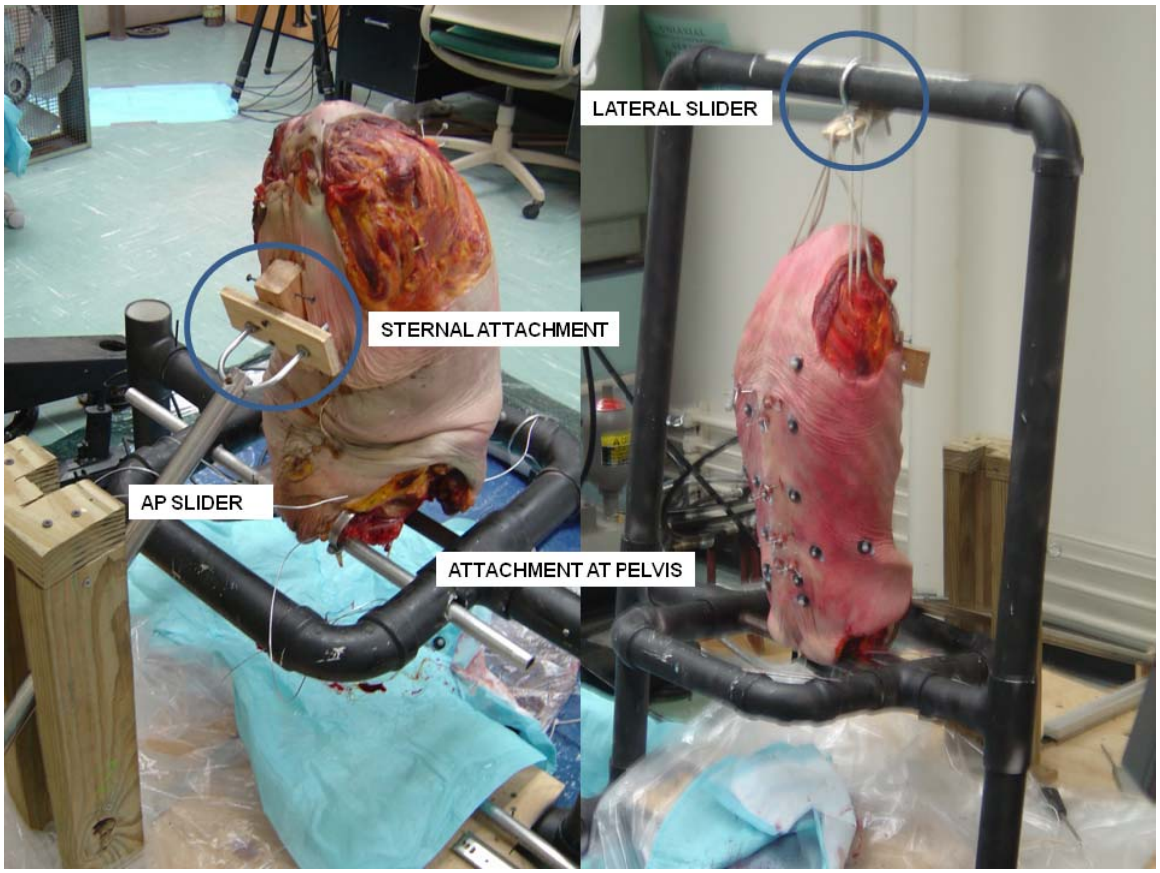


Figure 2.2. Testing Fixture

Testing fixture used for flexion (left) and lateral-bending (right).

2.2.6 Data Analysis

The three-dimensional coordinates of each marker were input into Visual 3D software (C-Motion Inc., Germantown, MD) to define the laboratory coordinate system, and local coordinate systems for the pelvis and each spinal level. The coordinates were smoothed using a 6 Hz Butterworth second order filter. Joint angles around all three axes were calculated for both the skin and pin marker triads by measuring the angles of the local coordinate systems at each level relative to the fixed laboratory coordinate system. The angles were calculated using 3 sequential rotations which allowed for the calculation of Cardan angles. The angles were zeroed at the beginning of the trials so that the origin of the motion was always zero. The differences between the skin marker and bone pin triads were calculated for all 15 trials of each motion.

2.3 Results

Representative 3D angle data for the skin-based markers and bone pin markers showing small (best-case) and large (worst-case) average differences between the skin marker and bone pin triad angles are shown (Figure 2.3). These data reveal close agreement in the angles measured by skin markers with measured bone pin angles for all tested motions, even in the trials with larger average angle differences.

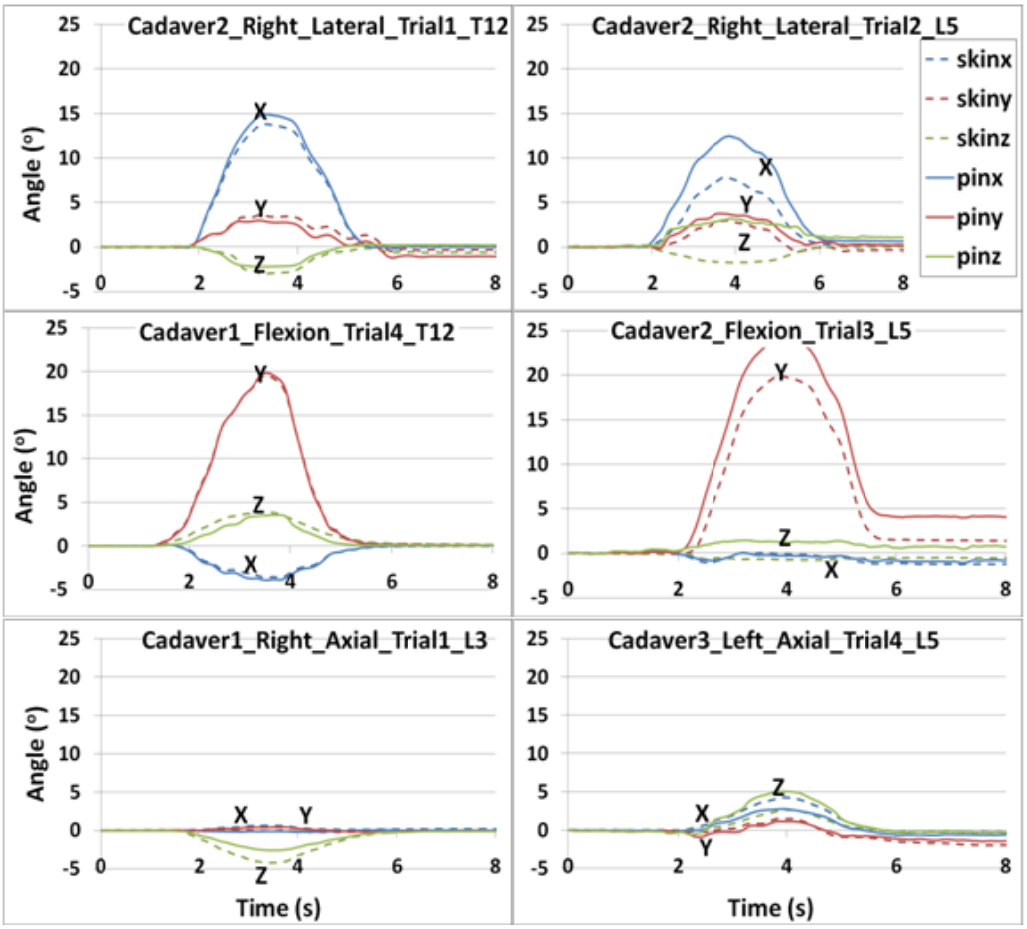


Figure 2.3. Vertebral Body Angular Data for Skin and Pin Markers

Representative 3D angle data for lateral-bending (x; top), flexion/extension (y; middle), and axial-rotation (z; bottom) of cadaver spines. Angle data illustrate differences between skin (dashed lines) and pin (solid lines) that are small (left) and large (right), i.e. best and worst case single trials.

A 3D plot of joint angles for a single cadaver flexion trial at the T12 level (Figure 2.4) reveals that curves for the skin and pin triads match closely with average differences of -0.01°, 0.08°, and 0.09° in lateral-bending, flexion and axial-rotation axes, respectively. Although the angles begin at zero, they do not return to zero because the torsos do not return to their exact starting position. The majority of flexion motion occurred about the y-axis with a ROM of approximately 18.0° (Figure 2.4). The ROM about the x and z axes were 1.3° and 1.8°, respectively.

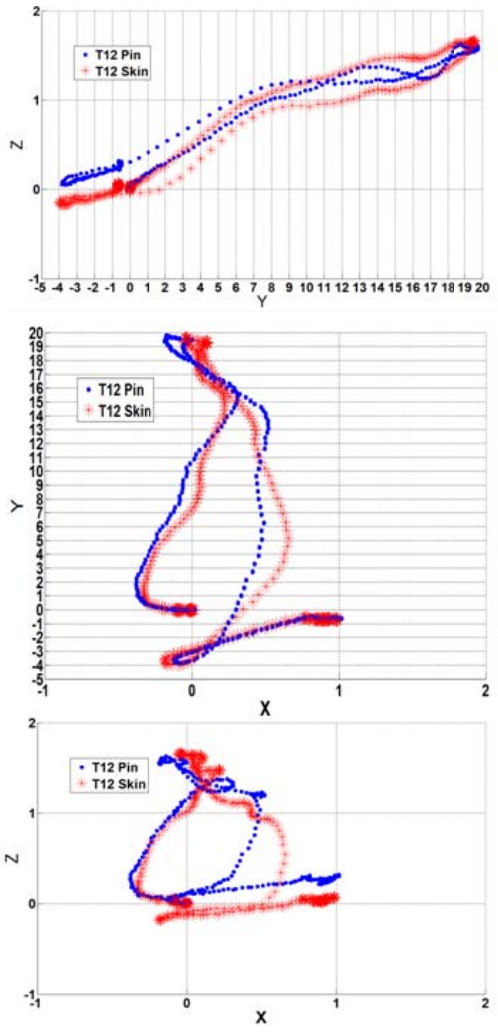


Figure 2.4. Vertebral Angle Data in 3 Planes.

Angle-Angle Plots showing representative angles (°) for skin triads (+, red) and pin triads (., blue) at the T12 level ina single cadaver torso during a flexion/extension trial.

The average range of motion of both skin and pin markers, the average differences between the two triads and the standard deviation are shown (Table 2.2). Mean differences in the angles given by the 2 pins at the T7 level vary from 0.1°-0.4° with the largest differences about the z-axis in left axial-rotation and left lateral-bending. Mean differences between skin marker and bone pin triads in all motions were ≤0.6° about the x- and y-axes in all cases except the L5 level in lateral-bending. The differences were less than 1.2° around the z-axis in all cases. In lateral-bending, the largest average angular difference was 1.1° at the L5 level around the x-axis, which was the main axis of motion. The overall ROM for the skin and pin triads were 7.0° and 10.4°, respectively. In flexion, the largest average angular difference measured by the two sets of markers was 1.3° at the L5 level around the y-axis (main axis of motion) with ranges of motion for the skin marker and bone pin triads of 14.2° and 16.4°. The largest average difference between triads in axial testing was 0.9° at the T7 level in the left axial testing around the z-axis which was again the main axis of motion. The overall ranges of motion for the skin marker and bone pin triads in this axial rotation trial were 8.6° and 11.6°.

Table 2.2. Comparison of Skin Marker and Bone Pin Marker Angles

Three-dimensional spinal angles (°) in lateral-bending, flexion/extension and axial-rotation (approximating to x, y and z directions) measured using skin (skin) and bone pin (pin) triads. Data entries in the table represent the average of 5 cycles for each of three torsos (15 total trials) presented for the range of motion (ROM), and average (Avg) and standard deviation (SD) of the differences between skin and pins. Negative numbers indicate that the pin markers angles were larger than skin marker angles.

Table 2.2. Comparison of Skin Marker and Bone Pin Marker Angles (continued)

Motion/Level	X				Y				Z			
Left_Lateral	skin ROM	pin ROM	Avg Angle Diff	SD	skin ROM	pin ROM	Avg Angle Diff	SD	skin ROM	pin ROM	Avg Angle Diff	SD
L5	8.1	9.9	0.8	0.2	8.6	7.6	0.5	0.1	2.6	3.5	0.9	0.4
L3	11.2	10.3	0.6	0.1	7.1	7.7	0.4	0.1	3.3	2.6	0.6	0.1
T12	14.0	13.8	0.3	0.0	4.5	5.3	0.5	0.2	2.5	3.1	0.6	0.2
T71	15.6	15.9	0.2	0.1	2.5	2.5	0.4	0.1	4.3	5.5	0.6	0.2
T72	15.6	15.8	0.2	0.1	2.5	2.2	0.3	0.2	4.3	5.5	0.5	0.1
T7pins	15.9	15.8	0.1	0.0	2.5	2.2	0.3	0.1	5.5	5.5	0.4	0.2
Right_Lateral	**X**				**Y**				**Z**			
L5	7.0	10.4	1.1	0.2	5.0	5.3	0.3	0.1	2.0	1.7	0.8	0.2
L3	11.2	10.6	0.3	0.1	4.7	5.2	0.3	0.1	2.1	1.3	0.5	0.1
T12	14.0	14.1	0.3	0.1	2.8	3.0	0.4	0.1	2.9	2.6	0.4	0.1
T71	16.6	16.7	0.1	0.0	3.1	2.7	0.3	0.0	2.5	3.5	0.3	0.0
T72	16.6	16.9	0.2	0.1	3.1	2.5	0.4	0.1	2.5	3.3	0.3	0.1
T7pins	16.7	16.9	0.1	0.0	2.7	2.5	0.2	0.1	3.5	3.3	0.2	0.0
Flexion	**X**				**Y**				**Z**			
L5	1.5	1.7	0.4	0.2	14.2	16.4	1.3	1.5	3.3	3.0	1.2	0.6
L3	2.0	1.7	0.3	0.1	15.3	15.7	0.5	0.1	3.1	2.2	0.9	0.2
T12	2.2	2.4	0.2	0.1	17.9	17.9	0.2	0.1	2.4	2.0	0.3	0.1
T71	2.9	2.7	0.3	0.1	19.0	18.5	0.2	0.1	1.8	1.0	0.4	0.1
T72	2.9	2.7	0.3	0.1	19.0	18.9	0.3	0.1	1.8	1.3	0.3	0.1
T7pins	2.7	2.7	0.1	0.0	18.5	18.9	0.3	0.1	1.0	1.3	0.2	0.1
Left_Axial	**X**				**Y**				**Z**			
L5	2.4	2.4	0.5	0.1	1.9	2.1	0.3	0.1	1.6	4.4	0.9	0.3
L3	2.5	2.2	0.4	0.1	1.8	1.9	0.3	0.1	3.6	3.7	0.3	0.1
T12	2.2	2.7	0.3	0.1	1.9	2.0	0.3	0.1	6.9	5.6	0.5	0.1
T71	1.9	1.5	0.3	0.0	1.8	1.6	0.3	0.1	8.6	11.6	0.9	0.1
T72	1.9	1.6	0.3	0.1	1.8	1.7	0.3	0.1	8.6	11.0	0.7	0.1
T7pins	1.5	1.6	0.1	0.0	1.8	1.8	0.2	0.1	11.6	11.0	0.4	0.1
Right_Axial	**X**				**Y**				**Z**			
L5	2.7	2.0	0.2	0.1	1.3	1.2	0.2	0.1	1.4	4.0	0.8	0.2
L3	2.6	1.9	0.3	0.1	0.9	1.2	0.2	0.0	3.7	3.4	0.3	0.1
T12	2.4	2.3	0.2	0.0	1.3	1.3	0.2	0.1	6.9	6.1	0.5	0.2
T71	1.3	1.2	0.1	0.1	1.9	1.6	0.2	0.1	8.6	11.4	0.9	0.5
T72	1.3	1.4	0.2	0.1	1.9	1.6	0.3	0.1	8.6	10.8	0.8	0.5
T7pins	1.2	1.4	0.1	0.0	1.6	1.6	0.2	0.1	11.4	10.8	0.2	0.1

Table 2.3. Measurement Error in the Lower Extremities During Gait

Average error between bone pin and skin marker angles in the lower body.

Authors	Anatomy	Lateral Bending	Flexion	Axial Rotation
Benoit et al.	Tibiofemoral	3.6°	2.5°	2.9°
Reinschmidt et al.	Tibiofemoral	2.1-2.8°	1.5-3.2°	2.1-5.3°
Reinschmidt et al.	Tibiocalcaneal	1.4-4.3°	2.2-4.4°	2.5-4.4°
Houck et al.	Tibiocalcaneal	1.5°	1.3°	1.0°
Andersen et al.	Tibiofemoral	2.1°	1.7°	2.7°
Nester et al.	Multiple bones in the Foot	2.3-3.9°	2.6-3.7°	1.9-5.1°

2.4 Discussion

This study showed that spinal motion can be noninvasively quantified by using skin markers. Differences between angles measured by the skin-based marker system and bone pin triads were typically less than 1°, even in ranges of motion that are larger than in normal gait. Previously reported ranges of motion in the spine for lateral-bending, flexion, and axial-rotation during gait are shown in Table 2.4.

Table 2.4. Reported Spinal Ranges of Motion During Gait

Frontal, sagittal, and transverse plane ranges of motion during human gait.

Study	Region	Frontal	Sagittal	Transverse
Current Study (not gait)	Thoracic/Lumbar	7°-17°	14°-19°	1.4°-11.4°
Crosbie (29)	Lumbar	9°	3.5°	4.5°
Crosbie (30)	Lower thoracic	7°	2.5°	4°
Feipel (41)	Lumbar	10°	5°	11°
Konz (82)	Lumbar	5.7°	4.1°	9.8°
Konz (82)	Thoracic	8.7°	5.7°	10.8°
Stokes (147)	Lower thoracic	4°	3°	5°
Vogt (158)	Lower thoracic	2.8°	2.4°	6.8°
Whittle (163)	Lumbar	7.55°	3.98°	8.34°

Differences between the angles measured at the different spinal levels and for different motions were observed. The differences based on spinal level could be attributed to the fixation of the pelvis or to an actual difference in the motion at the different spinal levels, but the differences can be observed using this measurement system. The maximum error in each trial occurred at the midpoint of the motion in maximum bending, flexion, or rotation.

The small differences (<0.5°) between the two pins implanted at the T7 level verify that this measurement technique provides equivalent motion data for two pins implanted into the same rigid body. The shape matching between the skin-based and bone pin triads was good in almost all trials in the five types of motion and at all spinal levels, even in trials where the average differences between the triads were largest. In general, the upper levels (T7 and T12) showed smaller average angle differences between the skin-based markers and bone pin triads than the lower levels (L3 and L5) with L5 consistently showing larger differences (Table 2.2). The larger differences observed at the L5 level could be due to effects of the fixation of the torso to the testing fixture at the pelvis. Average angle differences for flexion trials were usually smaller than those observed for lateral-bending and axial-rotation and differences measured

around the inferior-superior axis (z) were generally larger than around the other axes relative to the total range of motion for all motions tested (Table 2.2).

The thin torsos used in this study are an accurate representation of the population that suffers from spontaneous vertebral compression fractures. Multiple studies have shown a decreased body mass index leads to increased prevalence of these fractures with up to 40% of women over 50 with a BMI ≤18 kg/m² having at least one vertebral fracture (34,46,156). There is not a large mass of muscle or fat around the spine and there are many ligaments and fascia along the spine and between spinal levels connecting the skin to deeper structures that should allow for better results (less skin motion) than using skin markers in other areas of the body. The average differences found between bone pin angles and skin marker angles in other areas of the body are shown (Table 2.3).

IRB and human recruitment concerns limited this study to use of cadaver torsos. While simulated cadaver motions differ from actual motion in vivo, use of cadavers was justified given the comparative, not imitative, purpose of the study. Each trial recorded the motion of both sets of markers in a range that was similar to the motion that would be achieved in human subjects.

In conclusion, skin markers are a reasonable tool for quantifying the underlying vertebral motion and this finding enables subsequent studies to evaluate the hypothesis that biomechanical factors have a role in the etiology of low-energy vertebral fractures along with osteoporosis, to identify patients “at risk” for non-traumatic vertebral fracture, and to quantify the efficacy of various therapeutic means to correct abnormal postural or gait-related motion that may contribute to the incidence or severity of these fractures.

CHAPTER 3: SPINAL POSTURE AND GAIT-RELATED VERTEBRAL BODY MOTION IN NORMAL OLDER AND YOUNGER FEMALE SUBJECTS

3.1 Introduction

Osteoporotic vertebral fractures are a widespread health concern especially in thin, Caucasian women over 50 years old (34,46,156). An estimated 1.4 million vertebral fractures occurred worldwide in 2000 (48), and just over half of these (750,000) occurred in the United States (100). These fractures, commonly called "spontaneous" fractures, occur without a single large traumatic event and have been reported following relatively benign activities such as sneezing, stepping onto a curb, and bending or twisting the torso (112,128). Such fractures are characterized by collapse and loss of height (15-20%) of the vertebral body in the thoracic or lumbar regions (32,116,154). Spontaneous vertebral fractures can cause sudden and severe pain, abnormal curvature of the spine (possibly leading to deformity and Dowager's hump), immobility, reduced pulmonary function, difficulty performing activities of daily living, and may adversely affect forward vision which may in turn contribute to falls that lead to hip or forearm fractures (1,54,89,107,120).

Reduced load-bearing capability due to osteoporosis-related loss of bone mass is insufficient to explain the etiology of spontaneous vertebral fractures. Other factors must have a role in determining who suffers a fracture and when it occurs. Chief among these are biomechanical factors which have been theorized to interact with age-weakened osteoporotic bone to cause such fractures (23). Evaluation of this theory has been hindered by the lack of understanding regarding the motion of human vertebral bodies in vivo. Previous studies have attempted to noninvasively evaluate spinal kinematics using skin markers during walking (20,29,30,41,71,82,147,151,158,159,161,163,171), but for a variety of technical reasons, additional studies are needed to understand how vertebral bodies move during normal human gait. Therefore, the purpose of the current study was to quantify the gait-

related spinal motion parameters in older normal and younger normal subjects to determine whether there are age-related differences in these parameters.

3.2 Materials and Methods

3.2.1 Study Design

A cross-sectional clinical study was used to quantify the three-dimensional position and motion of vertebral bodies in females during postural (static standing) and gait (dynamic walking) related activities. Variables studied include: 1) age (older vs. younger), 2) vertebral level (thoracic and lumbar), 3) position and angle changes in three-dimensions, and 4) gait speed.

3.2.2 Subject Selection

Twenty-four healthy (no spinal fractures or surgeries) Caucasian females (Table 3.1) were recruited by word of mouth to participate in this Institutional Review Board approved study. Twelve older and twelve younger (mean age difference 32.1 years) subjects were recruited, provided informed consent and participated in this study.

Table 3.1. Subject Characteristic Data

Mean (± SD) Age, Height, Weight and Body Mass Index of Study Subjects

	Age(yrs)	Height(m)	Weight(kgs)	BMI
Older Normal	58.4(± 5.1)	1.67(± 0.1)	61.8(± 11.7)	22.0(± 2.3)
Younger Normal	26.39(± 3.7)	1.67(± 0.1)	60.0(± 7.3)	21.5(± 1.4)

3.2.3 Data Collection

Posterior/anterior (PA) and lateral radiographs of each subject's physiologically-loaded thoracic and lumbar spines were taken while they were standing upright. The radiograph view incorporated all vertebral bodies from T1 to L5. These images were

used to measure the: 1) lumbar lordosis angle, 2) thoracic kyphosis angle, and the 3) frontal Cobb angle (Figure 3.1).

The angle for lumbar lordosis was measured from the arc created by two intersecting lines parallel to the inferior endplate of the L5 vertebra and the superior endplate of the L1 vertebra. The angle for thoracic kyphosis was measured from the arc created by two intersecting lines parallel to the inferior endplate of the T12 vertebra and the superior endplate of the T5 vertebra. The T5 vertebral level was chosen as the superior limit because it was the highest level that could be clearly seen and measured in all subjects. The frontal Cobb angle was measured from the arc created by the inferior endplate of one vertebra to the superior endplate of the vertebra that creates the largest angle within a single lateral curve in the spine (24).

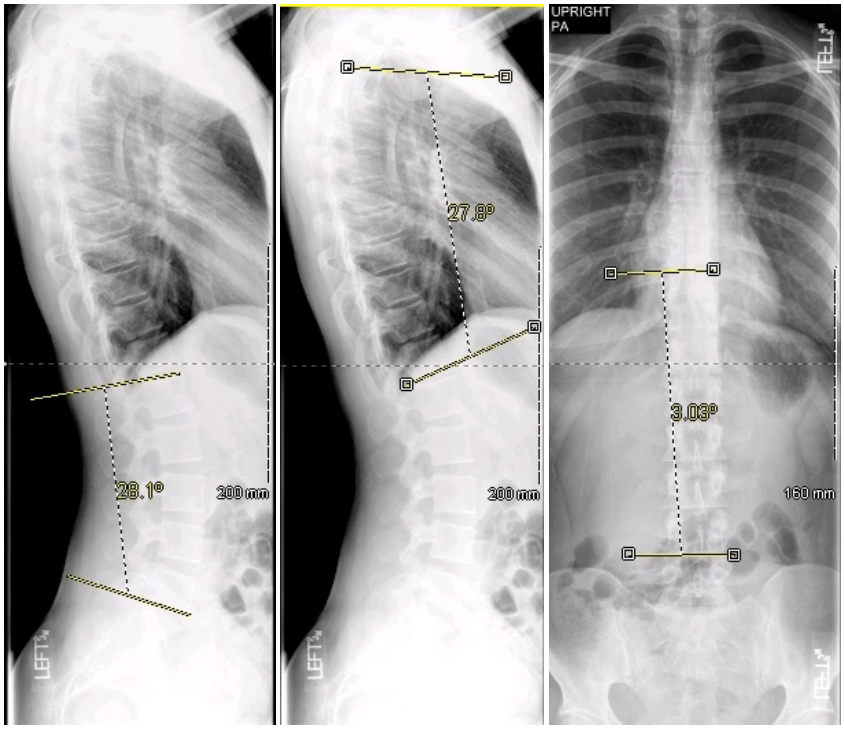


Figure 3.1 X-Ray Measurements

Sagittal and PA radiographic views of Subject 1 showing how lumbar lordosis (left), thoracic kyphosis (middle), and frontal Cobb angle (right) were measured.

During gait testing, each subject was asked to wear a specially prepared athletic shirt with an exposed back to allow markers to be placed on the skin over the spinous processes of the desired vertebral levels (T7, T10, T12, and L2). A total of 70 retroreflective 5mm diameter markers were placed on each subject. Markers at the spinal levels were placed in triads with an additional central marker. Additional markers were also placed on the legs, pelvis and arms to measure the motion of the upper and lower body (Figure 3.2). Calibration of the system was performed prior to each testing session. The calibration consisted of a static calibration in which an L-frame with markers that were at known positions was recorded by the motion analysis cameras. A wand calibration was then performed using a wand with 3 markers at known positions which was waved through the entire area that was to be calibrated. A static trial was taken of each subject while they were standing with their arms at their side on a fully instrumented dual-belt treadmill (Bertec Corporation, Columbus, OH) in order to aid in the definition of the rigid bodies. The subjects then walked on this treadmill for 4 trials at each of 3 selected speeds: 0.5 m/s, 0.7 m/s, and 0.9 m/s. These speeds were used based on the reported walking speeds of the elderly (60,80,83). Two-dimensional coordinates of the motion of each marker during treadmill gait were recorded by 15 high speed digital cameras and Cortex software (7 Eagle and 8 Eagle-4 cameras, Motion Analysis Corp, Santa Rosa, CA). The two-dimensional camera views of the markers were combined to determine the three-dimensional coordinates of each marker. Force data were collected from two plates embedded in the treadmill. The markers were identified and named and unknown markers were deleted before the trial was saved.

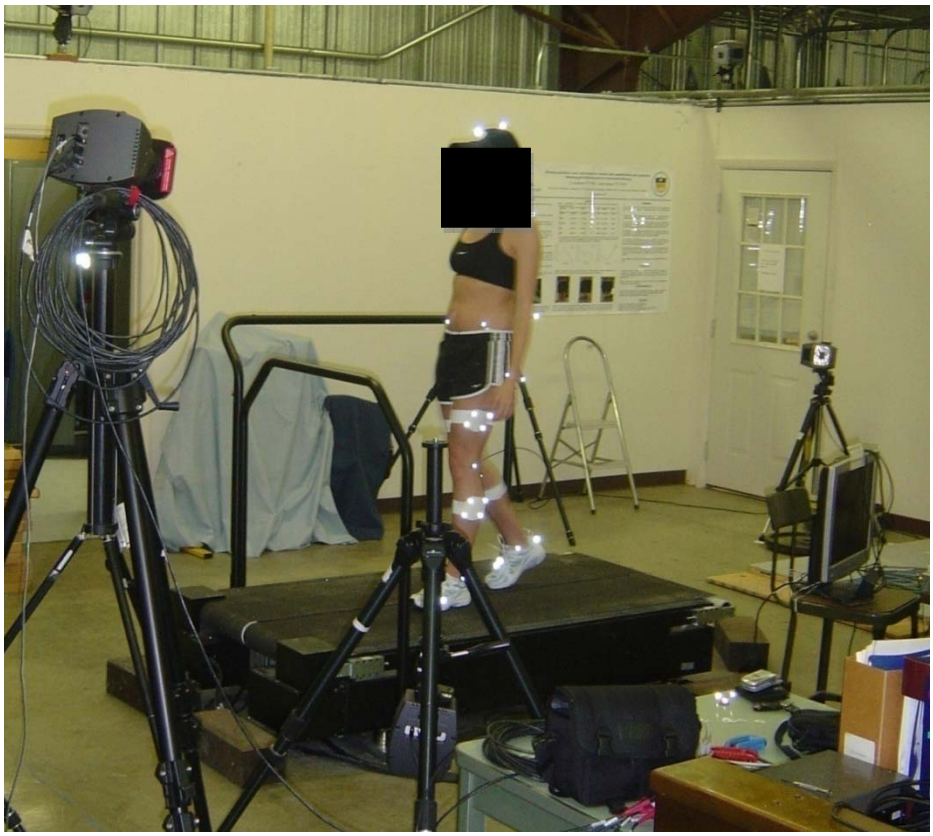


Figure 3.2. Subject on Treadmill

A subject walking on the treadmill instrumented with the full marker set.

3.2.4 Data Analyses

The three-dimensional coordinates of the markers were input into custom Matlab code (Mathworks, Natick, Massachusetts). Gaps in the marker coordinate data were joined by using a cubic spline. The coordinate and the force data were smoothed by using two low-pass (6 Hz and 50 Hz), Butterworth filters. The marker positions representing the pelvis and each of the 4 spinal levels were analyzed by using the custom Matlab code along with portions of KineMat Matlab code(123) to calculate the 3D angles of each vertebral level relative to the pelvis and the pelvis angles relative to the laboratory axis.

Left and right heel strikes were identified by using a threshold at 90N in the vertical force component of the ground reaction force. Each walking trial was divided into gait cycles starting with the first right heel strike and continuing until the next right

heel strike (Figure 3.3). A single trial included 3 - 8 gait cycles depending on the subject and walking speed. The angle data in three planes, measured for each spinal level relative to the pelvis and for each gait cycle, were normalized to 101 data points which allows for comparison as a percentage of gait cycle (Figure 3.3).

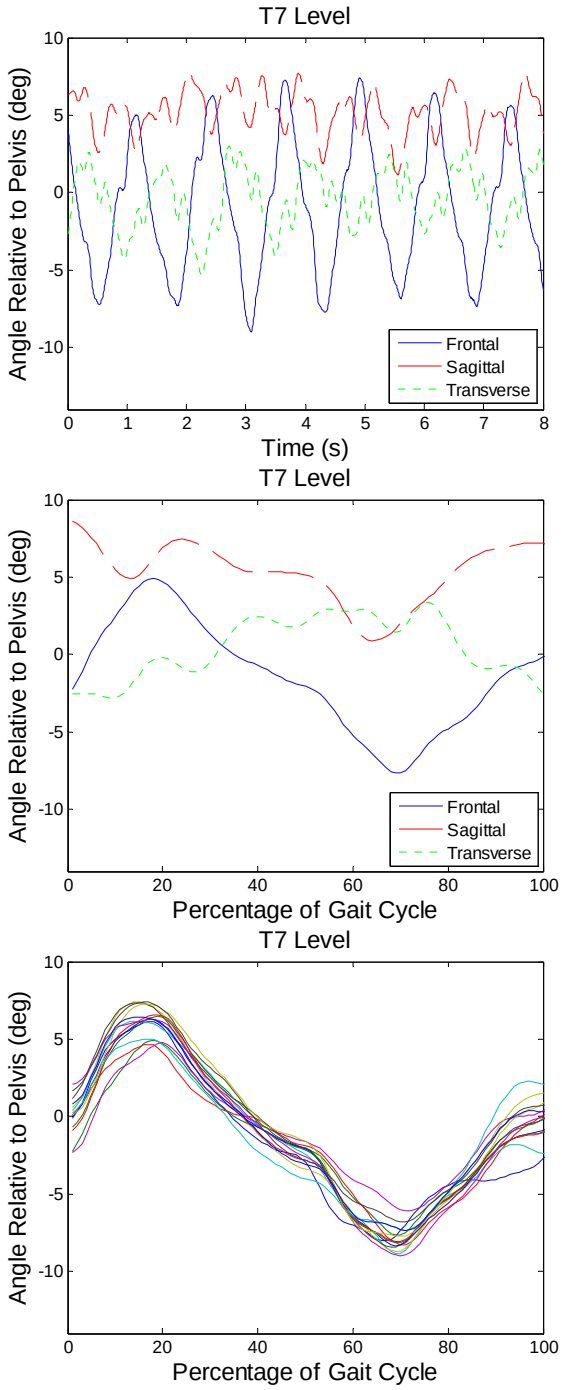


Figure 3.3. Angular Data Split by Gait Cycle

Angular data (relative to the pelvis) for the T7 level during a single walking trial in a normal younger subject (Top), the normalized angular data for a single gait cycle within this walking trial (Middle) and frontal plane angles for all gait cycles within this walking trial (Bottom).

The average angles from all gait cycles were calculated for: 1) each vertebral level and the pelvis, 2) each motion axis (flexion, lateral bending and axial rotation), and 3) each of the 3 gait-relevant speeds. A total of 45 (5 levels X 3 axes X 3 speeds) average curves were thus calculated per subject. The range for each average curve was calculated for each plane. As a measure of the step-to-step variation that occurred during gait, the standard deviation of each point along the average curve was calculated and then the mean of these 101 normalized points was determined (Figure 3.4). The range of motion and the variability (mean of the standard deviation) were compared by using an analysis of variance with the Scheffe's post-hoc correction.

3.3 Results

Significant differences in X-ray angles were not observed between the older and younger groups in lordosis or kyphosis angle, but the mean frontal Cobb angle was 31.7% greater in older subjects than younger subjects (Table 3.2).

Table 3.2. X-ray Angle Data

X-ray angle data for older and younger subjects.

	Lumbar Lordosis		Thoracic Kyphosis		Frontal Cobb Angle	
	Mean(±SD)	P-value	Mean (±SD)	P-value	Mean (±SD)	P-value
Older	39.7°±11.8°		29.0°±10.2°		8.2°±3.4°	
Younger	45.9°±6.9°	NS	26.8°±7.1°	NS	5.6°±2.1°	0.036

The largest ranges of motion in all spinal levels were observed in the frontal plane (lateral bending) and in the transverse plane (axial rotation) (Table 3.3). Motion observed in the sagittal plane (flexion) was 57.4% and 55.5% less (both $p<0.001$) than the motion observed in the frontal and transverse planes. Motion variability observed in the transverse and sagittal planes, as quantified by the standard deviations, was 23.2% and 21.0% greater than the motion variability observed in the frontal plane.

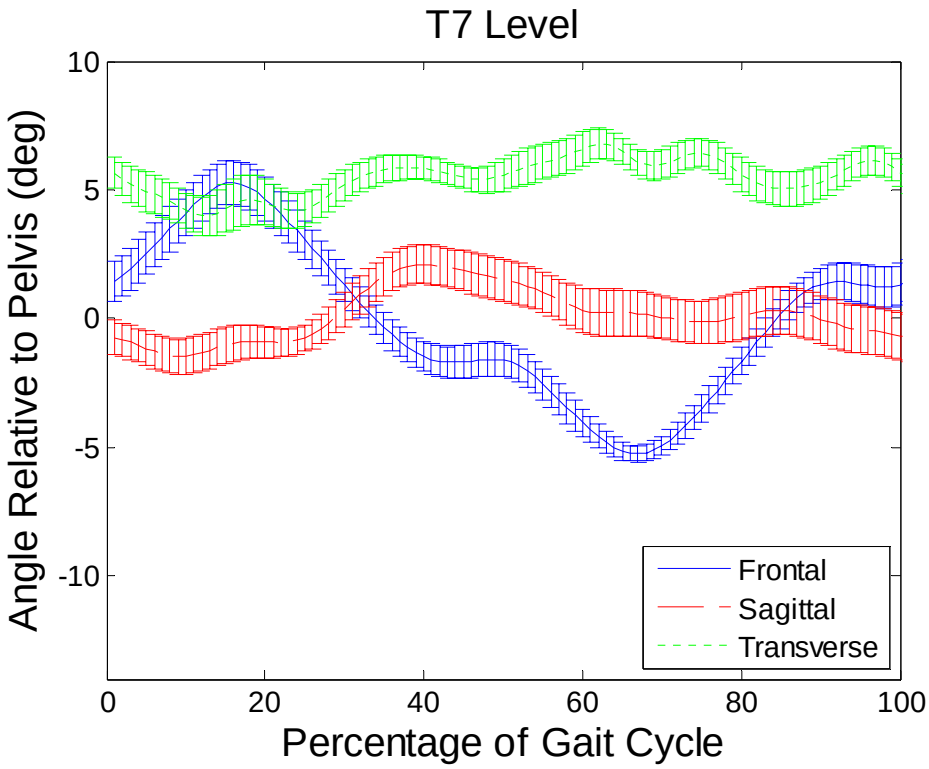


Figure 3.4. Mean Vertebral Angles

Mean angles (± SD) of the T7 level in the frontal, sagittal, and transverse plane obtained by averaging all gait cycles for a single subject walking at 0.9 m/s.

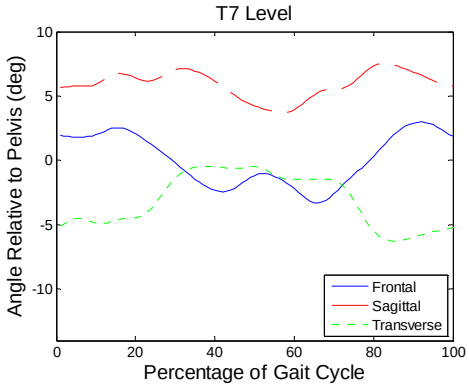
The relationship between walking speed and range of motion was determined for all subjects. As expected, faster speeds were associated with larger ranges of motion in lateral bending, flexion and axial rotation for all levels. The only significant differences in ranges of motion occurred between the fastest speed (0.9 m/s) and the slowest speed (0.5 m/s) with 13.3% higher ($p<.001$) ranges of motion at 0.9 m/s. There were no significant differences between the standard deviation values between the speeds.

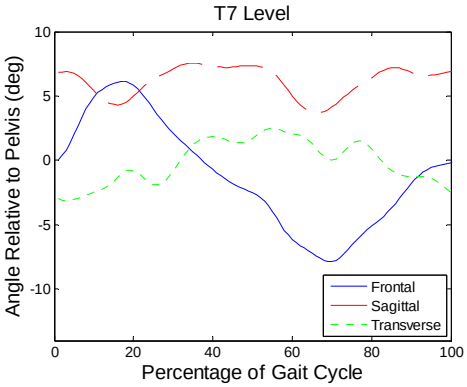
The older subject group had smaller ranges of vertebral body (all levels) and pelvis motion in the frontal plane for all gait speeds (Table 3.3). In the sagittal plane, the older subjects had a smaller range of motion at all spinal levels with similar pelvis ranges

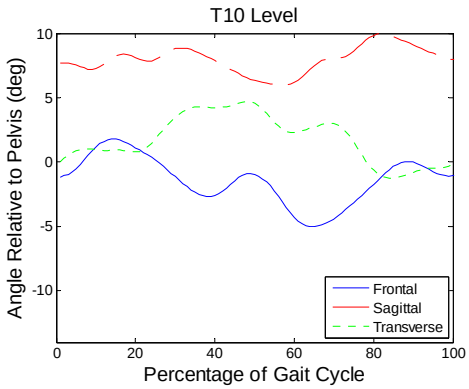
of motion. No significant differences were found between the groups in the transverse plane.

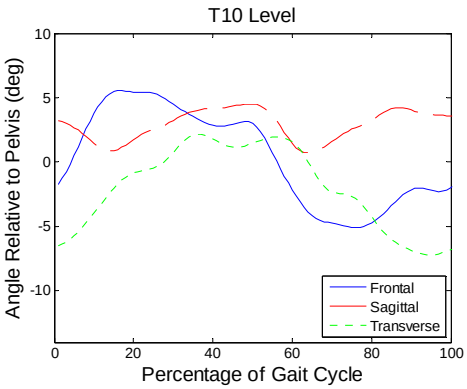
No significant differences in variability were found between the older and younger subjects with regard to frontal plane motion (Table 3.4). The majority of the statistically significant differences were seen in the sagittal plane, especially at the T12 and L2 levels, with the variability being smaller in the older subject group. Several significant differences in transverse plane standard deviation were also observed.

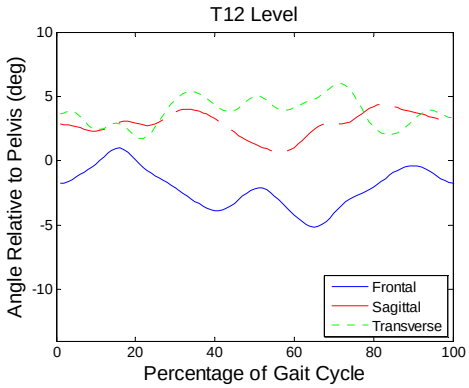
Gait patterns for a single older subject and a single younger subject (Figure 3.5) highlight the differences in shape between groups. The angular data in older subjects for frontal and sagittal plane motion was flatter than in the younger subjects showing the difference in ranges of motion. Maximum angular displacements in the frontal plane occur at left and right toe-off at roughly 20% and 70% of the gait cycle. The lateral angular displacement is towards the stance leg in the spinal levels and in the pelvis the opposite is true. The spinal segments are maximally extended at both heel strikes and maximally flexed at toe-offs in the sagittal plane. In the pelvis, maximum extension and flexion occurs slightly after the spinal levels. Transverse plane motion in the upper levels (T7 and T10) showed a maximum value at left heel strike and minima at right heel strikes. The lower levels (T12 and L2) show more oscillations. Maximum transverse plane angular displacement occurs slightly after toe off during leg swing in the pelvis.

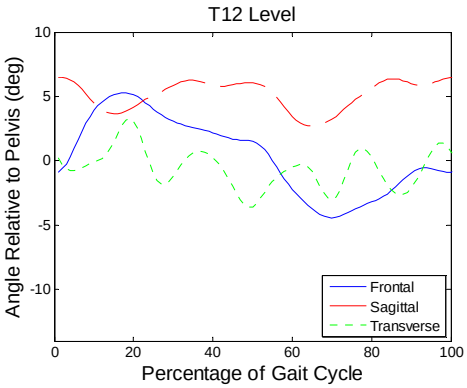

T7 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse

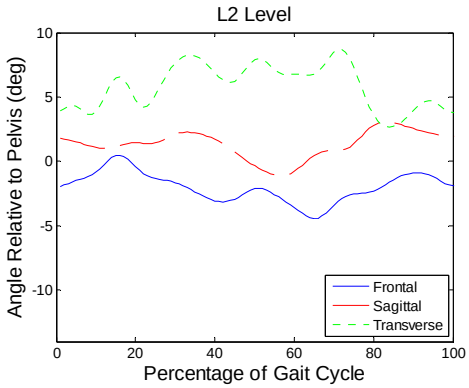

T7 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse

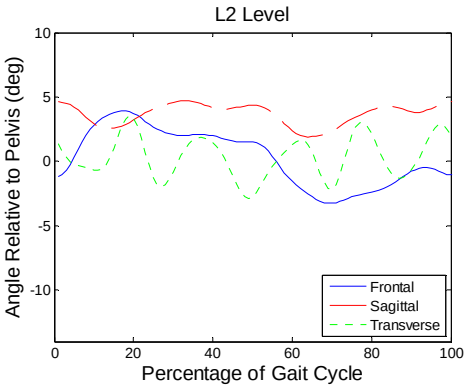

T10 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse


T10 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse


T12 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse


T12 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse


L2 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse


L2 Level
Angle Relative to Pelvis (deg)
Percentage of Gait Cycle
Frontal
Sagittal
Transverse

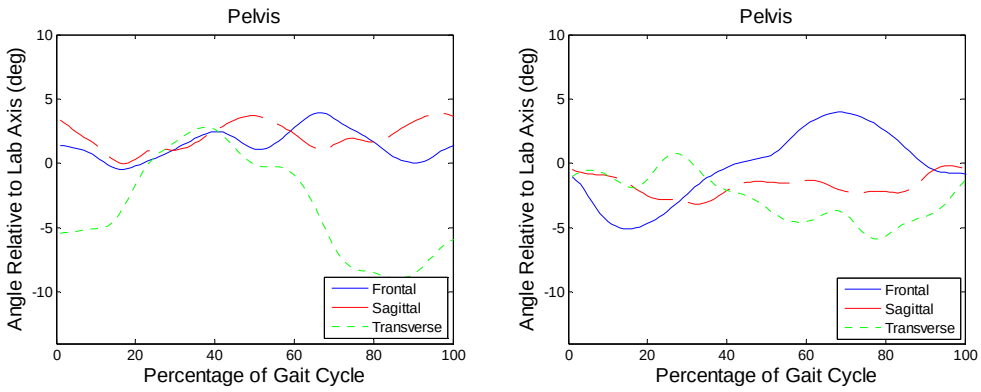


Figure 3.5. Representative Gait Patterns

Gait Patterns during gait from right heel strike to right heel strike for 1 older subject with smaller ranges of motion (left) and 1 younger subject with larger ranges of motion (right).

Table 3.3. Ranges of Motion During Gait

Ranges of motion in older and younger subjects separated by level, speed, and motion. Significant differences are shaded.

X-axis (Frontal Plane)	0.9 m/s			0.7 m/s		0.5 m/s	
		Mean ROM	P value	Mean ROM	P value	Mean ROM	P value
T7	Older	7.50	0.002	6.90	0.021	6.26	NS
	Younger	10.76		8.95		7.90	
T10	Older	7.06	NS	6.09	NS	5.75	NS
	Younger	8.82		7.80		6.65	
T12	Older	5.73	0.005	5.06	0.027	5.00	NS
	Younger	8.29		6.85		5.50	
L2	Older	4.75	0.049	4.05	NS	3.80	NS
	Younger	6.33		5.33		4.07	
Pelvis	Older	5.40	<0.001	4.95	<0.001	4.89	0.012
	Younger	9.00		8.00		6.82	
Y-axis (Sagittal Plane)	0.9 m/s			0.7 m/s		0.5 m/s	
		Mean ROM	P value	Mean ROM	P value	Mean ROM	P value
T7	Older	2.61	0.014	2.67	0.006	2.62	0.001
	Younger	3.72		3.93		3.94	
T10	Older	2.99	NS	2.97	0.012	2.75	0.004
	Younger	3.87		4.17		4.10	
T12	Older	2.68	NS	2.64	NS	2.46	0.018
	Younger	3.57		3.67		3.61	
L2	Older	2.81	NS	2.84	NS	2.56	0.022
	Younger	2.99		3.34		3.48	
Pelvis	Older	3.04	NS	2.79	NS	2.92	NS
	Younger	2.74		2.96		2.88	
Z-axis (Transverse Plane)	0.9 m/s			0.7 m/s		0.5 m/s	
		Mean ROM	P value	Mean ROM	P value	Mean ROM	P value
T7	Older	5.93	NS	5.42	NS	4.46	NS
	Younger	5.40		4.74		3.77	
T10	Older	8.35	NS	7.23	NS	5.79	NS
	Younger	7.99		7.24		6.39	
T12	Older	7.19	NS	7.04	NS	6.24	NS
	Younger	6.93		6.65		6.13	
L2	Older	7.80	NS	7.19	NS	7.04	NS
	Younger	7.22		7.62		7.11	
Pelvis	Older	6.37	NS	7.46	NS	8.42	NS
	Younger	6.96		7.64		8.80	

Table 3.4. Variability of Motion During Gait

Standard Deviation Values split by level, speed and motion. Significant differences are shaded.

X-axis (Frontal Plane)	0.9 m/s			0.7 m/s		0.5 m/s	
		Mean SD	P value	Mean SD	P value	Mean SD	P value
T7	Older	1.04		0.98		1.03	
	Younger	0.90	NS	0.98	NS	1.07	NS
T10	Older	1.14		1.06		1.09	
	Younger	1.05	NS	1.09	NS	1.10	NS
T12	Older	0.94		0.86		0.87	
	Younger	0.92	NS	0.94	NS	0.93	NS
L2	Older	0.69		0.63		0.63	
	Younger	0.64	NS	0.67	NS	0.65	NS
Pelvis	Older	0.68		0.62		0.63	
	Younger	0.54	NS	0.57	NS	0.67	NS
Y-axis (Sagittal Plane)	0.9 m/s			0.7 m/s		0.5 m/s	
		Mean SD	P value	Mean SD	P value	Mean SD	P value
T7	Older	1.13		1.04		1.10	
	Younger	1.40	NS	1.55	0.038	1.26	NS
T10	Older	1.14		1.10		1.20	
	Younger	1.05	NS	1.09	NS	1.10	NS
T12	Older	0.94		0.76		0.86	
	Younger	1.13	NS	1.29	0.004	1.16	NS
L2	Older	0.76		0.68		0.71	
	Younger	1.08	0.049	1.19	0.001	0.98	0.035
Pelvis	Older	0.91		0.89		0.97	
	Younger	0.85	NS	0.86	NS	0.85	NS
Z-axis (Transverse Plane)	0.9 m/s			0.7 m/s		0.5 m/s	
		Mean SD	P value	Mean SD	P value	Mean SD	P value
T7	Older	0.86		0.78		0.81	
	Younger	0.88	NS	0.75	NS	0.74	NS
T10	Older	1.03		0.86		0.94	
	Younger	1.23	NS	1.15	NS	1.11	NS
T12	Older	1.02		0.93		0.97	
	Younger	1.22	NS	1.37	0.016	1.20	NS
L2	Older	1.11		1.02		1.03	
	Younger	1.21	NS	1.25	NS	1.16	NS
Pelvis	Older	1.54		1.64		1.57	
	Younger	1.30	NS	1.31	0.033	1.35	NS

3.4 Discussion

This study is unique because unlike prior studies, the noninvasive measurement method used was validated with a prior cadaver study that employed bone pins. The present study showed that older subjects had smaller ranges of spinal motion in the frontal and sagittal planes and smaller variability in spinal motion in the sagittal plane as compared to younger subjects. Because the subjects were screened to exclude a history of back pain and or previous vertebral fractures, these differences in gait parameters are considered attributable to the difference in age in these female subjects.

The previously validated (146) noninvasive method presently used for measuring spinal motion showed similar gait patterns and ranges of motion in normal subjects as those reported elsewhere (29,30,41,71,82,158,159,161,163). The decreased spinal ranges of motion with advancing age as measured during walking shown presently match the findings of others (36,38,135,153) who also showed decreases in spinal range of motion with increasing age. It is worthy to note; however, that these previous studies measured spinal motion only during active and passive lateral bending, flexion and axial rotation during standing or sitting as opposed to gait. There was a linear decrease in ranges of motion with age in these studies. Specifically, motion decreases were reported to be: 40% in flexion, 76% in extension, 43% in lateral bending and no change in axial rotation (153). These results are comparable to those of the current study which found reductions in frontal and sagittal plane ranges of motion with age but no differences in axial rotation.

There is a paucity of data reporting spinal motion variability. Previous studies (20,149) examine the repeatability of multiple test sessions (day to day variability) rather than the stride-to-stride variability during a single testing session as was measured in the current study. The decreased variability in spinal motion seen in the older subjects is the opposite trend as found in lower body variability in other studies (9,14,17,53,58,59,69,73,78,114,145). The parameters that were investigated in these studies include: stride length, stride time, knee angles and step width, and in general an increase in variability with age was found. The opposite trends in spinal and lower

extremity variability could be due to the fact that the older subjects may have a loss of strength, mobility or balance with age that would cause them to walk more cautiously with less upper body motion even while variability in the lower extremities could be higher.

The decreased ranges of motion and variability in spinal motion may mean that spinal loading in older subjects is not as uniformly distributed across bony load bearing surfaces as well as it is in younger subjects. Less variable loading could lead to damage accumulation in a portion of the vertebrae that is faster than the rate of bone remodeling which decreases with age (136). The greater resulting loads could lead to fatigue fractures occurring with fewer loading cycles than would occur in younger subjects that more uniformly distribute such loads. Older and younger subjects may have different time-dependent loading profiles. The time during which gait-related axial loads are applied to each "micro area" of a given vertebrae may be longer for older adults due to the reduced range and variability of motion. Smaller ranges of motion in the frontal and sagittal planes in older subjects would likely load the central and anterior portions of the vertebral body more than the lateral and posterior regions. The time dependent differences in bone loading may alter the response of bone remodeling. This may help explain the kyphotic-related anterior portion vertebral body collapse commonly observed in spinal fractures in post-menopausal women. The greater variability in younger subjects could distribute the loading pattern across a larger area of bone and therefore may lessen the damage to the bone allowing for repair through remodeling. The fracture mechanism may be more dependent on the time loading history of regions of the vertebral body rather than the number of cycles.

The difference (2.6°) in mean frontal Cobb angle observed between the older and younger subjects was significant, but none of the subjects had frontal Cobb angles that were in the range of clinical scoliosis requiring treatment. Frontal Cobb angles less than 10° are not a clinical concern and angles less than 20° generally do not require any treatment (52). The increased frontal Cobb angles seen in older subjects, while not of short-term clinical concern, may be biomechanically significant. The higher frontal Cobb

angles seen in the older subjects may correlate with the dynamic testing results. The increased curvature of the spine may contribute to the decreased range of motion and decreased spinal motion variability in the frontal plane during treadmill gait. A retrospective study found that frontal Cobb angles were significantly higher in subjects who suffered a vertebral fracture than in non-fractured subjects (102). Scoliotic subjects were also found to have less variable spine motion than normal subjects in the study by Chan et al. showing the association of increased frontal Cobb angle with reduced spinal variability (20).

This technique was limited to use in subjects with a body mass index (BMI) less than approximately 25 because larger subjects have too much soft tissue between the skin markers and the vertebrae. This is not a major limitation because spontaneous vertebral fractures are most prevalent in those without a surfeit of adipose tissue. Also, motion measurement was limited to every other spinal level due to the size of the marker triads. This was not a major fault due to the limited motion of the spinal segments and the ability to interpolate motion from adjoining segments.

In conclusion, differences in spinal posture and spinal kinematics were shown between older and younger female subjects. These postural and kinematic differences may in turn cause altered spatial and temporal spinal loading. Such altered loading, in the presence of bone already weakened by material loss due to osteoporosis, may be related to the mechanism of spontaneous vertebral body fractures. Confirmation of this hypothesis in a future prospective study may enable the development of simple interventions such as orthotics, gait retraining, or targeted muscle strengthening to prevent or retard the onset of future spontaneous vertebral fractures.

CHAPTER 4: OVERALL DISCUSSION AND CONCLUSIONS

4.1 Key Findings

A model for the noninvasive measurement of spinal motion using skin markers was developed and validated in a cadaver study. Skin markers were able to accurately quantify vertebral body motion, relative to bone pin markers, in cadaver torsos. Average differences between angles measured by the skin-based marker system and bone pin triads were less than 1° in all planes. The ranges of motion tested were larger than the ranges seen in normal gait. These results showed that this skin-based marker system was a reasonable tool for evaluating spinal motion and therefore enabled the use of the system to compare motion in human subjects.

The validated model was able to measure differences in spinal motion parameters between older normal and younger normal female subjects. The older subject group had reduced ranges of motion in the frontal and sagittal planes of up to 30.9% and 33.5% respectively. Another parameter of interest was variability in spinal motion quantified by mean standard deviation. The differences in variability were most evident in the sagittal plane where the older group had reduced variability of up to 42.9% as compared to the younger group.

4.2 Discussion

The cadaver study was an important step to show what differences in spinal motion parameters are meaningful in human subjects. The cadaver testing protocol tested ranges of motion that were larger than those seen during gait in human subjects. The differences between skin-based markers and bone pin markers were generally still less than 0.6° in the frontal and sagittal planes and less than 0.9° in the transverse plane. Human subject testing showed significant differences in spinal range of motion between the older subjects and the younger subjects in the frontal and sagittal planes. Differences over 0.6° in the frontal and sagittal planes and 0.9° in the transverse plane

are significantly different meaning that there is a difference that is not due solely to skin sliding motion. A significance difference in an experiment does not always indicate that a clinical significance exists however. A difference of less than a degree would be hard to consistently measure clinically. Differences over 1° between subjects would likely be required for clinical observation. This level may need to be higher (1.5°) in the transverse plane due to the larger error associated with using skin markers in that plane.

There were significant differences between older and younger subjects in 8 out of 15 cases in mean frontal plane range of motion (4/5 levels at the 0.9 m/s; 3/5 levels at 0.7 m/s; 1/5 level at 0.5 m/s). The differences in mean frontal plane range of motion varied from 1.8° to 3.3° in the significantly different cases. All of these differences in the frontal plane should indicate a clinically meaningful difference since they are over 1°. The smallest mean difference that was significantly different between older and younger subjects in frontal plane motion (1.8°) was 1.2° larger than the 0.6° difference between skin and bon-pin markers in the frontal plane seen in the cadaver study. When including the non-significant differences in frontal plane range motion between the older and younger subjects, only three cases, all at the slowest speed, were below 1° of difference.

Significant differences in sagittal plane ranges of motion between the older and younger subjects were observed in 7 of 15 cases (1/5 at 0.9 m/s; 2/5 at 0.7 m/s; 4/5 at 0.5 m/s). The mean differences in sagittal plane motion varied from 0.9° to 1.35° at the significantly different levels. These differences would still be clinically relevant but the differences would not be as clear as in the frontal plane.

Significant differences in frontal and sagittal plane ranges of motion were shown between the older and younger subject groups. The magnitude of difference be clinically relevant has also been discussed. A discussion of why these differences exist is the next logical step. The changes the body undergoes during aging are likely the cause of the altered spinal kinematics between subjects. These changes can be evident in many areas including: posture, intervertebral disc degradation, facet joint degradation,

ligament laxity, and muscular weakening. The viscoelastic nature of the soft tissues and bone were key in explaining the differences that occur over time.

The intervertebral disc changes with age as the proteoglycan content of the disc decreases, leading to a decrease in the water content (6). A corresponding increase in collagen fibers, with type I fibers replacing type II fibers, can lead to increased cross-linking which changes the characteristics of the disc (33,155). The loss of hydration of the nucleus of the disc can cause the annulus fibrosis to bear more compressive load (37). The blood supply to the vertebral endplate also decreases with age which can allow for microtears to accumulate. It becomes increasingly difficult with age for the disc to remove and repair accumulated degradation products (130).

The facet joints can also be greatly affected by aging. The facet joints help stabilize and control the motion of the spine and are in almost constant motion. This leads to a thinning of the cartilage in the facet joint which can lead to bony growth and enlargement of the joint causing osteoarthritis and pain. The relationship between the intervertebral disc and the facet joints is an important factor in determining the stability of the spine. It is generally thought that disc degeneration begins before facet joint degeneration since the disc carries a larger percentage of the compressive load (44).

Ligament laxity can also influence the rate of degeneration of the disc and facet joints (8). Laxity in the spinal ligaments can be due to injuries throughout the life of the subject, normal aging, and genetics (8). Subjects with increased laxity exhibit increased degeneration of the intervertebral disc which can be due to the increased forces seen in the surrounding joints. The loss of stability in the ligaments could lead to altered motion patterns.

Changes in muscle properties with age may also contribute to vertebral fractures. Aging muscles lose size and strength and therefore must generate higher forces just to keep the body in correct posture. The back extensor muscles are required to generate even more force if other factors of aging described previously have caused the upper body to be at a forward leaning angle compared to what the ideal posture would be. Increased muscle forces are transferred to the vertebral bone and disc. These

greater muscle forces can lead to degeneration of the disc and vertebral body causing altered kinematics and loading of the disc and vertebral body.

4.3 Limitations

The method of spinal motion measurement was validated in three cadaver torsos. Ideally, more than three torsos would help to further verify the results, but the results were consistent across the three cadavers. Cadavers were chosen for this study due to IRB issues and the difficulty in recruiting human subjects for a bone pin study. Cadavers cannot account for active muscle contraction which could change how the skin moved in relation to the underlying bone. Fresh cadavers were used to minimize the differences in skin properties between live subjects and the cadaver skin. Soft tissues and muscle contractions in humans may actually reduce the motion if the skin over the vertebrae because the muscles would be active in normal posture, walking, and in any flexion/extension, lateral bend, or axial rotation. The ranges of motion that were tested in the cadavers were larger than what would be seen in normal walking so the amount of motion of the skin should at least be equivalent to what would be seen in live subjects during walking.

The subject groups in human subject testing were older normal and younger normal females. These groups allowed for comparison of spinal motion parameters over time, but they do not give direct answers to the question of why do certain people suffer a fracture. The original research plan included subjects with a previous vertebral fracture, but finding subjects with a previous fracture proved to be difficult. Another concern was that any differences that were found in a fractured group versus a normal group could have existed pre or post-fracture. The actual cause of the differences in kinematics could be investigated with future studies described in a following section.

Marker triads could only be placed at every other spinal level due to their size. Preliminary testing determined that the markers needed to be 4 cm apart to avoid camera errors that can cause the angle data to have large errors. The motion of the

entire spine can still be measured at every other level, and future camera technology may allow for every level to be measured.

This measurement technique is limited to use in thin subjects with a BMI of 25 or less. This is not an issue in this study since the majority of people who suffer fractures are thin females as shown in Figure 4.1. Other research studies using the technique would also be limited to thin subjects only unless a separate validation study was done in human subjects or cadavers with higher BMI.

Table 4.1. The Effect of BMI on Fracture Rate

Prevalence of vertebral deformities and incident vertebral fracture risk as a function of BMI as reported in two separate studies.

BMI	Prevalence of Vertebral Deformity (46)		BMI Quartile	Incident Vertebral Fracture Risk (156)
< 18	40%		**Lowest**	2
19-24	35%		**2**	1.5
25-29	19%		**3**	1.2
> 30	16%		**Highest**	1

A longitudinal study would be able to provide more definitive results as to whether gait parameters were actually predictive of vertebral fractures. A longitudinal study is difficult to complete however due to cost, difficulty in keeping subjects enrolled, and the time required.

4.4 Clinical Implications

If it is shown that people can be placed in an “at-risk for fracture” group based on noninvasive measures, then the efficacy of clinical treatment methods can be examined. Intervention could begin after an initial fracture in an effort to prevent further fracture, or the intervention can be applied before fracture to subjects that are determined to be at-risk based on their medical history or their gait parameters (i.e. reduced range of motion and reduced variability). These interventions could be non-

surgical in subjects where a fracture has not occurred or surgical treatments may be required in patients with a previous fracture.

Surgical treatments for previous fracture include kyphoplasty and vertebroplasty. Both involve injection of bone cement into a collapsed vertebra to prevent further collapse. Kyphoplasty also includes insertion of a balloon that is inflated within the vertebral body prior to the injection of bone cement in order to restore some of the lost vertebral height. The purpose of these procedures is to restore more normal posture and motion in the spine which will also aid in restoring normal loading of the vertebrae.

Non-surgical countermeasures would include simple measures such as medications (anti-inflammatory, antiresorptive, or anabolic), vitamins, weight training, flexibility or balance training, orthotics, gait or postural training, bracing, tai chi, targeted strengthening of the back muscles and fall prevention training.

Exercising has been shown to offer a variety of benefits in older women including better posture, increased mobility, less pain and dependence on pain medication, better reported quality of life, and better balance with less fear of falling.

Targeted strength training of back flexor and extensor muscles can be beneficial in preventing vertebral fractures (77,90,139,140). Sinaki et al. (139) examined bone mineral density, spine radiographs, back extensor strength, biochemical marker values, and levels of activity in 50 post-menopausal women at baseline, after 2 years of treatment, and 8 years after the end of treatment. Twenty-seven women completed a resistive back strengthening exercise over the two year treatment and 23 women were control subjects. Mean back extensor strength in the exercise group was 39.4 kg at baseline, 66.8 kg after 2 years of strengthening, and 32.9 kg 8 years after the end of treatment compared to 36.9 kg at baseline, 49.0 kg at 2 years, and 26.9 kg at 10 years in the control group. The bone mineral density of the exercise group was significantly higher in the exercise group at the 10 year mark even after no difference was seen at baseline and at 2 years. The relative risk of fracture was shown to be 2.7 times greater in the control group with 4.3% of vertebrae being fractured in the control group versus

1.6% in the back exercise group. Similar experiments were performed by Kemmler et al. and Lord et al. and the results showed increased bone density, increased trunk muscle strength and decreased pain in the exercise group versus the control group.

Balance that has been lost progressively over time in the elderly can be at least partially restored by balance training (84,95,143,165) or the use of orthotics (105,148). Balance and proprioception were quantified using functional tests of single and double-leg stances, with eyes open and eyes closed. All of these balance training and orthotic studies showed significantly better balance after treatment. Better balance can lead to better posture and more range of motion in gait which could aid in varying the loading patterns of the vertebrae.

Gait training goes hand-in-hand with balance training and can also increase range of motion during walking as well as increase stride length. These factors could help prevent spinal fractures by increasing the area of the vertebrae that is subjected to loading.

The efficacy of all of these interventions could be quantified using the same noninvasive measurement system to compare spinal motion parameters before and after treatment.

4.5 Conclusions

Based on the results of this research it can be concluded that:

I. Skin based markers can provide a reasonable tool for quantification of vertebral body motion in subjects with BMI ≤ 25.

II. This validated method enables new comparisons of spinal motion parameters in previously unexamined applications.

III. Older female subjects showed decreased ranges of motion in the frontal and sagittal planes during treadmill gait compared to younger subjects.

IV. Older female subjects showed less variability in spinal motion (especially in the sagittal plane) during treadmill gait when compared to younger subjects.

The data from this study leads to a new hypothesis: Spinal fractures may be associated with abnormal static or dynamic gait-related loading patterns which may interact with age weakened osteoporotic bone to cause such fractures. More specifically, decreases in the range and variability of vertebral body motion in older subjects may increase gait-related stresses on the vertebral bodies. These increased stresses may lead to bone damage from a fatigue mechanism and this may contribute to the cause of spontaneous non-traumatic vertebral fractures. Three factors are vital to understanding this possible fracture mechanism: 1) the amplitude of the force, 2) the location of the force application, and 3) the time duration of the loading. Evaluation of this hypothesis awaits a prospective study, but if proven, the etiology of these fractures could be advanced and new cost-effective preventative strategies may emerge. This validated noninvasive measurement system that can detect gait parameter differences between subject groups can also be applied to a number of situations beyond spine fractures including implant performance comparison (disc replacement or facet replacement), surgical procedure assessment (i.e. comparison of kyphoplasty vs. vertebroplasty), balance and falling studies and would allow for extension of previous studies that stopped analysis at the hips.

4.6 Future Research

A logical continuation to the present research is a longitudinal study. The height, weight, and BMI of all subjects would be matched as in the previous study. The ages of the subjects would be matched as well as risk factors for fracture and a measure of bone quality such as bone mineral density. A detailed questionnaire would be completed by the subjects to determine hereditary factors and other risk fractures that could preclude them from the study. Initial x-rays would also be taken to ensure that no previous vertebral fractures existed.

Spinal motion parameters during gait would be collected for a large number of subjects at an initial time point. These subjects would be followed over time possibly with follow up visits to collect gait data at set time points. The subjects who suffer vertebral fractures can be placed in a fractured group. The spinal motion parameters of this group could then be compared to the matched subjects who have not suffered a fracture. The parameters identified in the previous study, spinal ranges of motion and variability of spinal motion, would be examined initially, but differences may also be evident in parameters that have not yet been identified. Other studies could be completed in the event that a longitudinal study is not possible. These studies would have less time requirements and less associated cost.

Mechanical testing of vertebrae could be performed using the known kinematics from the completed study. Two loading profiles could be used to investigate the effect of the time loading history on the vertebrae. One loading profile would represent the younger group seen in the completed study in that the loading would be distributed over larger ranges of motion in more variable patterns. The other loading profile would represent the older group and have the loading distributed over a smaller range of motion and in less variable patterns. All of the vertebrae should have similar shape, size, and bone mineral densities so that the differences in the number of cycles before fracture would be due to the loading profiles only. The null hypothesis would be that the two groups of vertebrae would show no difference in the number of loading cycles necessary to cause a fracture.

The known kinematics from the completed study could also be applied to an inverse dynamics model. A method such as Kane's dynamics (81) could be used to calculate the contact forces between the vertebral levels. The null hypothesis would that no loading differences would exist between the older and younger subjects. CT scans or MRI would be needed to increase the accuracy of the model. This would allow for accurate geometry of the vertebrae and the addition of muscle forces, but it would also add to the cost of the study and increase the radiation exposure of the subjects.

Another study that could be completed in a shorter time span and at less cost than a longitudinal study would be a comparison of previously fractured subjects versus normal subjects. Differences in spinal parameters could be determined using the same methods as in the completed study. The effect of age could be removed by testing subjects in the same age range while keeping height, weight, and BMI equivalent between groups. The null hypothesis would be that there are no differences in the spinal posture and gait-related motion between fractured and non-fractured subjects. This study will be useful because the results of this study could be compared to the completed study to see if the differences show up in the same parameters: 1) frontal Cobb angle, 2) spinal ranges of motion, and 3) variability of spinal motion or whether the changes are evident in other spinal motion parameters. Comparing non-fractured subjects to fractured subjects would not be as useful without the knowledge of the differences between the older and younger subjects. The cause of any differences could have existed prior to fracture or could be a result of the fracture.

The effect of various treatments on frontal Cobb angle, spinal ranges of motion, and variability could be examined in another study. This would allow for the quantification of the efficacy of interventions that may be able to delay the onset of or prevent vertebral fractures by restoring more normal motion. Two subject groups with equivalent ages, height, weights, BMI, and BMD would be used with one being a control group and the other being a treatment group. More groups could be tested based on the number of treatment methods to be investigated. The null hypothesis would be that the treatment group and the control group would have equivalent changes in spinal parameters at a follow up data collection after a certain treatment period.

APPENDIX A: FULL LIST OF MARKERS

Cadaver Study: 51 refelective markers total

1. Top marker on the skin marker triad at the T7 spinal level
2. Bottom left marker on the skin marker triad at the T7 spinal level
3. Bottom right marker on the skin marker triad at the T7 spinal level
4. Central marker on the skin marker triad at the T7 spinal level
5. Top marker on the skin marker triad at the T12 spinal level
6. Bottom left marker on the skin marker triad at the T12 spinal level
7. Bottom right marker on the skin marker triad at the T12 spinal level
8. Central marker on the skin marker triad at the T12 spinal level
9. Top marker on the skin marker triad at the L3 spinal level
10. Bottom left marker on the skin marker triad at the L3 spinal level
11. Bottom right marker on the skin marker triad at the L3 spinal level
12. Central marker on the skin marker triad at the L3 spinal level
13. Top marker on the skin marker triad at the L5 spinal level
14. Bottom left marker on the skin marker triad at the L5 spinal level
15. Bottom right marker on the skin marker triad at the L5 spinal level
16. Central marker on the skin marker triad at the L5 spinal level
17. Top marker on the first bone pin triad at the T7 spinal level
18. Bottom left marker on the first bone pin triad at the T7 spinal level
19. Bottom right marker on the first bone pin triad at the T7 spinal level
20. Central Marker on the first bone pin triad at the T7 spinal level
21. Top marker on the second bone pin triad at the T7 spinal level
22. Bottom left marker on the second bone pin triad at the T7 spinal level
23. Bottom right marker on the second bone pin triad at the T7 spinal level
24. Central Marker on the second bone pin triad at the T7 spinal level
25. Top marker on the bone pin triad at the T12 spinal level
26. Bottom left marker on the bone pin triad at the T12 spinal level

27. Bottom right marker on the bone pin triad at the T12 spinal level
28. Central Marker on the bone pin triad at the T12 spinal level
29. Top marker on the bone pin triad at the L3 spinal level
30. Bottom left marker on the bone pin triad at the L3 spinal level
31. Bottom right marker on the bone pin triad at the L3 spinal level
32. Central Marker on the bone pin triad at the L3 spinal level
33. Top marker on the bone pin triad at the L5 spinal level
34. Bottom left marker on the bone pin triad at the L5 spinal level
35. Bottom right marker on the bone pin triad at the L5 spinal level
36. Central Marker on the bone pin triad at the L5 spinal level
37. Left transverse process marker at the T7 level
38. Right transverse process marker at the T7 level
39. Left transverse process marker at the T12 level
40. Right transverse process marker at the T12 level
41. Left transverse process marker at the L3 level
42. Right transverse process marker at the L3 level
43. Left transverse process marker at the L5 level
44. Right transverse process marker at the L5 level
45. Left posterior superior iliac spine marker (pelvis)
46. Right posterior superior iliac spine marker (pelvis)
47. Left ischial tuberosity marker (pelvis)
48. Right ischial tuberosity marker (pelvis)
49. Upper offset marker on the back - used to differentiate left from right
50. Middle offset marker on the back - used to differentiate left from right
51. Lower offset marker on the back - used to differentiate left from right

Human Subject Study: 70 reflective markers total

1. Top marker on the skin marker triad at the T7 spinal level
2. Bottom left marker on the skin marker triad at the T7 spinal level

3. Bottom right marker on the skin marker triad at the T7 spinal level
4. Central marker on the skin marker triad at the T7 spinal level
5. Top marker on the skin marker triad at the T10 spinal level
6. Bottom left marker on the skin marker triad at the T10 spinal level
7. Bottom right marker on the skin marker triad at the T10 spinal level
8. Central marker on the skin marker triad at the T10 spinal level
9. Top marker on the skin marker triad at the T12 spinal level
10. Bottom left marker on the skin marker triad at the T12 spinal level
11. Bottom right marker on the skin marker triad at the T12 spinal level
12. Central marker on the skin marker triad at the T12 spinal level
13. Top marker on the skin marker triad at the L5 spinal level
14. Bottom left marker on the skin marker triad at the L5 spinal level
15. Bottom right marker on the skin marker triad at the L5 spinal level
16. Central marker on the skin marker triad at the L5 spinal level
17. Back offset marker
18. Left anterior superior iliac spine (pelvis)
19. Right anterior superior iliac spine (pelvis)
20. Sacral marker (pelvis)
21. Anterior hat marker (head)
22. Superior hat marker (head)
23. Posterior hat marker (head)
24. Lateral hat marker (head)
25. Left lateral condyle of the humerus (elbow)
26. Left medial condyle of the humerus (elbow)
27. Right lateral condyle of the humerus (elbow)
28. Left medial condyle of the humerus (elbow)
29. Left ulnar styloid (wrist)
30. Left radial styloid (wrist)
31. Right ulnar styloid (wrist)

32. Right radial styloid (wrist)
33. Left upper arm marker
34. Left lower arm marker
35. Right upper arm marker
36. Right lower arm marker
37. Left acromion process (shoulder)
38. Right acromion process (shoulder)
39. Left toe marker (foot)
40. Left heel marker (foot)
41. Left laces marker (foot)
42. Left lateral marker (foot)
43. Right toe marker (foot)
44. Right heel marker (foot)
45. Right laves market (foot)
46. Right lateral marker (foot)
47. Left lateral maleolus (ankle)
48. Left medial maleolus (ankle)
49. Right lateral maleolus (ankle)
50. Right medial maleolus (ankle)
51. Left lateral condyle of the femur (knee)
52. Left medial condyle of the femur (knee)
53. Right lateral condyle of the femur (knee)
54. Right medial condyle of the femur (knee)
55. Superior left shank marker
56. Anterior left shank marker
57. Inferior left shank marker
58. Posterior left shank marker
59. Superior right shank marker
60. Anterior right shank marker

61. Inferior right shank marker
62. Posterior right shank marker
63. Superior left thigh marker
64. Anterior left thigh marker
65. Inferior left thigh marker
66. Posterior left thigh marker
67. Superior right thigh marker
68. Anterior right thigh marker
69. Inferior right thigh marker
70. Posterior right thigh marker

APPENDIX B: CALCULATION OF JOINT ANGLES

The methods for calculating the joint angles of the spinal levels relative to the pelvis during the human subject testing will be described in detail in this section.

Three-dimensional coordinates of the markers were calculated from the multiple two-dimensional camera views using the known calibration data and the focal lengths of the cameras. The markers were identified using the Cortex software (Motion Analysis, Santa Rosa, CA) and then the coordinate data was input into a custom Matlab code. The rigid bodies that were modeled include: head, left and right upper arms, left and right lower arms, pelvis, T7 spinal level, T10 spinal level, T12 spinal level, L2 spinal level, left and right thigh, left and right shank, and left and right foot. The bodies of the most interest in this study were the spinal levels and the pelvis. Each rigid body had a minimum of three markers associated with it. Each marker has 3 columns of coordinates (x,y, and z coordinates) with 300 and 800 rows respectively for static and dynamic trials (3 and 8 seconds recorded at 100 Hz). Any gaps in the input coordinates were filled using cubic spline interpolation and the interp1 function in Matlab. The coordinate data was then smoothed using a 6 Hz Butterworth second order low-pass filter.

Force plate data was also collected and input into the custom Matlab code where it was filtered with a 50 Hz Butterworth second order low-pass filter. This force plate data was used to identify left and right heel strikes and toe offs. This was done using a threshold in the vertical component of the ground reaction force. Passing the threshold in the positive direction created the heel strikes and crossing going in the negative direction created the toe offs. The dual-belted treadmill allowed for separation of left and right using the separate signals from the two plates.

The joint angles reported were cardan angles with a rotation sequence of ZXY. This sequence was chosen with the largest motions first, with Z being axial rotation, X being lateral bending, and Y being flexion/extension. The local coordinate systems of the spinal levels were set up to have anatomically relevant axes. The triads were arranged

with two markers along the line of the vertebral endplate and one marker superior to the midpoint of the first two markers. The Y axis was defined as the axis between the two markers along the endplate, the Z axis was the axis created in the direction of the superior marker, and the X axis was perpendicular to the other axes with the positive direction being forward on the treadmill. The coordinates of both bodies during the static trial and during a gait trial are needed to get the angle of one body relative to another throughout the motion. The coordinates are used as inputs into the cardan function taken from the Kinemat Matlab code. The cardan checks the sizes of the matrices and then calls on the soder function. The soder function outputs the 4 x 4 transformation matrix using singular value decomposition. Another function, rzxysolve, uses the input of the transformation matrix to calculate the joint angles from the direction cosine matrix. These angles are the angles reported for each vertebral level.

The methods in the cadaver study were similar in that cardan angles were calculated. These calculations were done using a commercially available software (Visual 3D, C-Motion, Inc., Germantown, MD) and no custom Matlab code was used.

An example of the calculation of cardan angles follows. An x,y,z sequence of rotations will result in the rotation matrix R. R is a combination of the three independent rotation matrices associated with the 3 rotations (R_x , R_y , and R_z).

$$[R]=(R_x][R_y][R_z]$$

where:

$$[R_x] = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos\alpha & \sin\alpha \\ 0 & -\sin\alpha & \cos\alpha \end{pmatrix}$$

$$[R_y] = \begin{pmatrix} \cos\beta & 0 & -\sin\beta \\ 0 & 1 & 0 \\ \sin\beta & 0 & \cos\beta \end{pmatrix}$$

and $$[R_z] = \begin{pmatrix} \cos\gamma & \sin\gamma & 0 \\ -\sin\gamma & \cos\gamma & 0 \\ 0 & 0 & 1 \end{pmatrix}$$

The combined rotations matrix is then:

$$[R] = \begin{pmatrix} \cos\beta\cos\gamma & \cos\gamma\sin\beta\sin\alpha + \sin\gamma\cos\alpha & \sin\gamma\sin\alpha - \cos\gamma\sin\beta\cos\alpha \\ -\sin\gamma\cos\beta & \cos\alpha\cos\gamma - \sin\alpha\sin\beta\sin\gamma & \sin\gamma\sin\beta\cos\alpha + \cos\gamma\sin\alpha \\ \sin\beta & -\cos\beta\sin\alpha & \cos\alpha\cos\beta \end{pmatrix}$$

The elements in [R] represent the relative orientation of one coordinate system relative to another (spinal level relative to the pelvis). The cardan angles are calculated directly from the [R] matrix and are the three independent projection angles α , β , and γ.

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EDUCATION

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PROFESSIONAL POSITIONS

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HONORS

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PEER-REVIEWED PUBLICATIONS

- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Mullineaux DR, Shapiro R. Spinal Posture and Gait-Related Vertebral Body Motion in Normal Older and Younger Female Subjects. Manuscript to be submitted to Spine. April, 2011.
- **Stinton SK**, Marrs BH, Shaffer WO, Cassidy RC, Pienkowski DD, Shapiro R. Biomechanical Comparison of 5 Lumbo-pelvic Fixation Designs. Manuscript submitted to Spine. January, 2011.
- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Mullineaux DR, Shapiro R. Development and Validation of a Noninvasive Spinal Motion Measurement System. Manuscript submitted to Spine. January, 2011.

CONFERENCE PRESENTATIONS

- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Mullineaux DR, Shapiro R. Development and Validation of a Noninvasive Spinal Motion Measurement System. Electronic Poster presented at the North American Spine Society Annual

meeting. Orlando, FL. October, 2010.

- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Mullineaux DR, Shapiro R. A Novel Biomechanical Approach for Preventing Spinal Disorders. Poster presentation at the KSEF-KSTC annual conference. Lexington, KY. April, 2010.
- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Mullineaux DR, Shapiro R. Development and Validation of a Non-Invasive Spinal Motion Measurement System. Podium Presentation at the annual meeting of the American Society of Biomechanics. State College, PA. August, 2009
- Cunningham TJ, Mullineaux DR, **Stinton SK**. Intra-Limb Joint Coupling Patterns During the Use of Three Lower Extremity Exercise Machines. Podium Presentation at the annual International Conference on Biomechanics in Sports. Limerick, Ireland. July, 2009.
- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Mullineaux DR, Shapiro R. A Biomechanical Approach for preventing Spinal Disorders. Poster presentation at the KSEF-KSTC annual conference. Louisville, KY. April, 2009.
- **Stinton SK**, Shaffer WO, Cassidy RC, Pienkowski DD, Shapiro R. Lumbopelvic fixation: biomechanical support for the concept of a pelvic foundation in spinal surgery. Poster presented at the annual meeting of the International Society for the Study of the Lumbar Spine. Bergen, Norway. June, 2006.
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- **Stinton SK**, Komistek RD, Mahfouz MM. 3D motions of normal and degenerative lumbar spine patients. Scientific Poster presented at the annual meeting of the American Academy of Orthopaedic Surgeons. Washington DC. February, 2005.
- Fei L, **Stinton SK**, Komistek RD, Mahfouz MM. Mathematical model of detrimental effects of cervical spine fusion. Poster presented at the annual meeting of the International Society for Technology in Arthroplasty. Rome, Italy. July, 2004.
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