TRUNK LEAN IN CONTROL AND OSTEOARTHRITIC GAIT

by

HEATHER SARAH LINLEY

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Abstract

Trunk lean over the stance limb during gait has been linked to a reduction in the knee adduction moment, which is associated with joint loading. Differences were examined in knee adduction moments and frontal plane trunk lean during gait between subjects with knee osteoarthritis and a control group of healthy adults. Additionally, subject variability in human motion data presents a challenge to researchers when trying to detect differences between subject groups. The individual differences in neutral posture between subjects is a source of variation in joint angles. A method was developed using principal component analysis (PCA) to objectively reduce this inter subject variability.

Gait analysis was performed on 80 subjects (40 osteoarthritis). Models were developed to define lateral thoracic tilt, as well as pelvic tilt. The trunk and pelvis frontal plane angles were used to describe trunk lean and pelvic tilt. Angles were calculated across the stance phase of gait. We analyzed the data, (i) by extracting discrete parameters (mean and peak) waveform values, and (ii) using principal component analysis (PCA) to extract shape and magnitude differences between the waveforms.

Osteoarthritis (OA) subjects had a higher knee adduction moment than the control group (α=0.05). Although the discrete parameters for trunk lean did not show differences between groups, PCA did detect characteristic waveform differences between the control and osteoarthritis groups. The data show that subjects display similar waveform shapes, however waveforms vary in magnitude, suggesting a variation in posture between subjects. The results from the PCA reveal that the first PC, which captures the most variation in the data, represents this variation in magnitude. The second PC describes a significant difference in range of motion between the subject groups.

Subjects with knee OA were found to have a different range of motion of their pelvis and trunk than control subjects. These changes are consistent with a strategy to lower the knee adduction moment. As an alternative to conventional subjective methods, PCA should be employed to reduce inter subject variability in order to ensure objective analysis in human motion waveform data.

Co-Authorship

Heather Linley was the primary author of all chapters of this thesis. Chapter 2 is presented as a manuscript which was submitted to the journal *Clinical Biomechanics* in June 2009. Co-authors of this manuscript are Dr. Elsie Culham, Dr. Elizabeth Sled, and Dr. Kevin Deluzio, The study was designed in collaboration with all authors, and data collection was done by Dr. Elizabeth Sled. The data analysis and the manuscript were completed by Heather Linley. The manuscript was then edited by all co-authors.

Chapter 3 is also presented as a manuscript which has been prepared for submission as a brief communication. The co-author of this manuscript is Dr. Kevin Deluzio. The study was designed by both authors. The analysis and writing of the manuscript was done by Heather Linley, and editing was done by both authors.

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Chapter 1

1. Introduction

1.1. Knee Osteoarthritis

The prevalence of osteoarthritis increases with age, making it a primary health concern for the future as our population distribution sees a rise in the number of older adults (Felson *et al.*, 2000). Although it has been reported that osteoarthritis is the most common reason for lower limb joint replacement surgeries (Felson *et al.*, 2000), only a small fraction of patients diagnosed with osteoarthritis actually receive operative treatment for the disease (Buckwalter *et al.*, 2001). Instead, it has been suggested that alternatives to medical or pharmacological treatment for joint protection can be employed by the patient themselves (Brant, 1998).

Recent studies have proposed the possible adoption of gait alterations or retraining as potential methods of reducing the pain caused by and progression of knee osteoarthritis (Chang *et al.*, 2005; Hunt *et al.*, 2008; Mundermann *et al.*, 2008).

1.2. Gait Analysis

Gait analysis is performed in order to quantify the patterns of human motion. By comparing the gait of healthy control subjects to the gait of those with identified pathologies, we can begin to gain a better understanding of the mechanics associated with degenerative diseases. Several variables are used to characterize individual gait patterns. These may be spatiotemporal such as gait speed, cadence, and stride length, or kinematic

and kinetic factors such as joint forces, moments, and angles.[1] Gait is cyclic, and can be divided into two phases; stance and swing. The loading of the lower limb joints are characterized when the limb of interest is in stance. This comprises approximately 60% of the full gait cycle and is defined between heel strike and toe off of the limb of interest (Astephen and Deluzio, 2009).

1.3. Knee OA and the knee adduction moment

The external knee adduction moment, or the torque tending to bend the knee toward bowleggedness, has been shown to be a good surrogate for evaluating the loading in the medial compartment of the knee joint (Figure 1.1) (Hurwitz *et al.*, 1998; Baliunas *et al.*, 2002; Hurwitz *et al.*, 2002). It is related to the distribution of forces between the medial and lateral compartments of the knee (Baliunas *et al.*, 2002) and has been shown to be related to disease severity and possibly progression (Chang *et al.*, 2005; Sharma *et al.*, 1998). Additionally, in 2008, Astephen *et al.* found that both severe and moderate knee OA subjects display higher mid-stance external knee adduction moments compared to controls. It should be noted that it has been suggested that the increased knee adduction moment found in subjects with OA may not be the initial cause of the pathology, but rather, the effect of morphologic changes in the diseased joint (Mundermann *et al.*, 2004).

[1] A detailed methodology of gait analysis is provided in Appendix A.

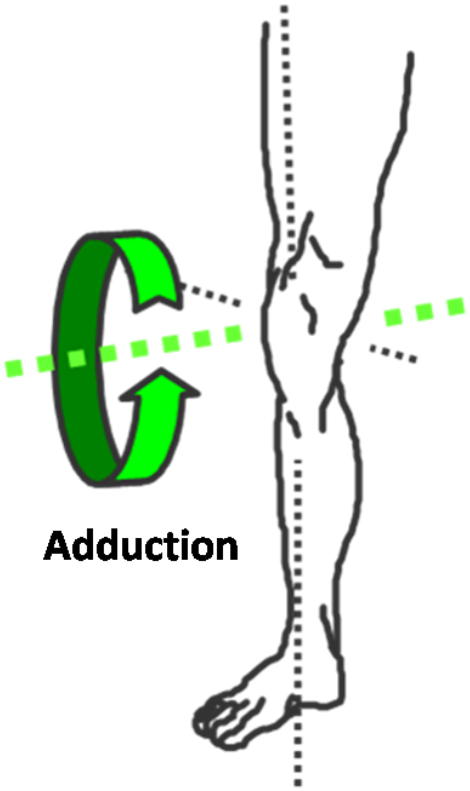

Figure 1.1: Knee adduction moment. The adduction moment acts about the anterior-posterior axis and causes a tendency toward bowleggedness.

1.4. Adduction moment compensations (gait alterations) in OA subjects

The external knee adduction moment can be increased or reduced by altering gait patterns to adjust the load distribution in the joint. Several gait alterations have been suggested as either a strategy that patients with knee OA make on their own, or as possible non invasive interventions which could be used clinically. Walking with the toes pointed out has been found to reduce the adduction moment in the knee by displacing the centre of pressure laterally (Andrews *et al.*, 1996; Chang *et al.*, 2007; Hunt *et al.*, 2008; Hurwitz *et al.*, 2002; Teichtahl *et al.*, 2006). Wedged insoles used on the lateral side of the shoe have also been found to reduce the medial compartment loading in the knee

joint, although these results were more effective in subjects with mild OA (Brandt, 1998; Sasaki and Yasuda, 1987; Yasuda and Sasaki, 1987). Additionally, some patients have been found to decrease their external knee adduction moment by reducing their self-selected gait speed; however this was most prevalent in those with less severe knee OA, and was found to be patient specific (Mundermann *et al.*, 2004).

1.5. Hip adduction moment, connection with knee OA

In 2005 Mundermann *et al.* hypothesized that subjects with knee OA experience increased load rates at the ankle and hip as well as at the knee during gait, particularly in the frontal plane. They found significant differences in first and second peak adduction moments between subjects with more severe OA and controls with the OA subjects displaying lower peak values. Another study looked to identify gait variables which distinguish between control and OA subjects with both moderate and severe OA (Astephen *et al.*, 2008). They found that both knee OA groups had reduced peak and first peak hip adduction moments compared to controls. Chang *et al.* (2005) proposed that patients display increased internal hip abduction moments as a method to protect against medial compartment OA progression. This hypothesis however was not quantifiably verified. In 2008 it was reported that when walking with increased medio-lateral trunk sway, healthy subjects displayed significantly decreased first peak hip adduction moments compared to their typical gait (Mundermann *et al.*, 2008). Additionally, it has been most recently found that although hip abductor strength is inherently one of the contributing factors to influence the moments about the hip in the frontal plane, gait

velocity and subject mass were found to explain the greatest amount of variability in the adduction moment at the hip (Rutherford and Hubley-Kozey, 2009). These studies suggest that knee OA does in fact have an effect on the other joints of the lower limb, and is specifically tied in with gait alterations at the hip joint.

1.6. Trunk lean, impact on knee adduction moment

It has been speculated by researchers that increasing the frontal plane lateral sway in the trunk is an effective method for reducing the external adduction moment in the knee (Hunt *et al.*, 2008; Mundermann *et al.*, 2005; Mundermann *et al.*, 2008) (Figure 1.2). Mundermann *et al.* (2008) found that increased lateral trunk lean does reduce the knee adduction moment in healthy subjects supporting results by Hunt *et al.* (2008) who found that lateral trunk lean correlated highly with the first and second peak knee adduction moments compared to other kinematic variables in OA subjects. Although Hunt *et al.* (2008) speculated that this gait alteration primarily originates in the thorax alone due to the relative low pelvic obliquity (or, frontal plane pelvic tilt) angles, it should be noted that the angles of trunk lean reported by the authors are of a similar magnitude to their reported angles of pelvic tilt. Therefore it is possible that the pelvic tilt in the frontal plane should be examined in conjunction with thoracic tilt.

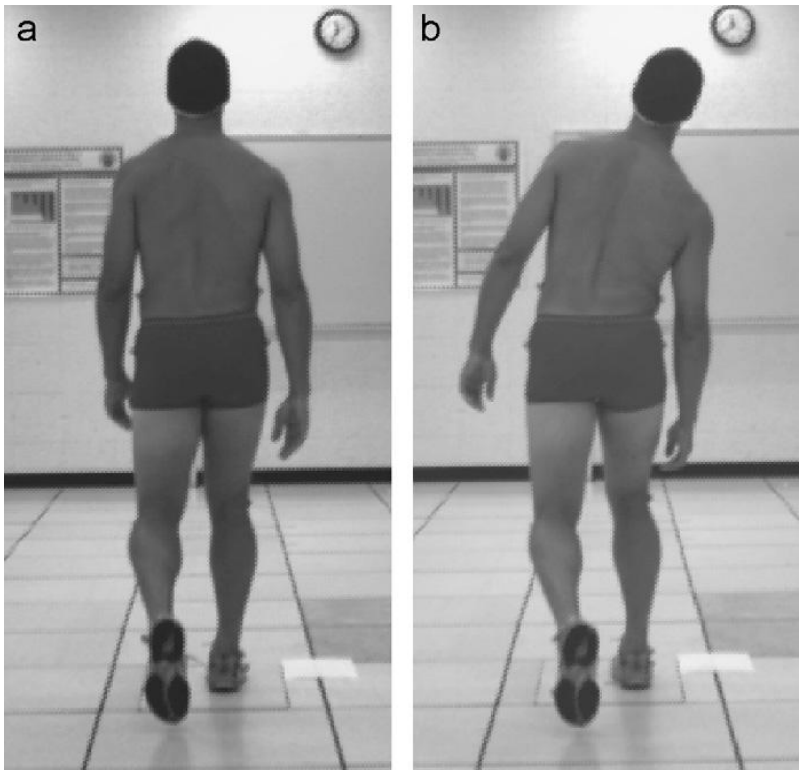

Figure 1.2: Normal and increased medio-lateral trunk sway (Mundermann *et al.*, 2008). A healthy subject walking with (a) normal trunk sway and (b) purposefully increased trunk sway.

1.7. Measurement variations in the trunk and pelvis

Measurement techniques vary among authors when attempting to track the motion of the thorax, or trunk, and the pelvis. Several studies track the lumbar spine and pelvis as separate segments. Rigid clusters of reflective markers are often attached to the skin near the level of the thoracolumbar junction using wands in order to track the motion of the lumbar spine (Taylor *et al.*, 1999; Whittle and Levine, 1999; Schache *et al.*, 2002). These wands are also often used to track the motion of the pelvis when placed over the sacrum (Taylor *et al.*, 1999; Whittle and Levine, 1999) or alternatively, rigid body coordinate systems have been defined in the pelvis by directly capturing the motion of bony landmarks with reflective markers (Schache *et al.*, 2002). The motions of these lumbar

and pelvis segments are then tracked either with respect to one another or with respect to a defined laboratory coordinate system.

Recently, there has been a trend toward tracking the full thorax as a rigid body. One study examining trunk lean motion during gait in subjects with knee OA defined trunk lean as the angle between the line of cervical vertebrae C7 to sacral vertebrae S1 and a perpendicular line to the laboratory floor throughout the gait cycle (Tanaka *et al.*, 2008). Both Mundermann *et al.* (2008) and Hunt *et al.* (2008) defined the angle of lateral trunk lean by using the frontal plane projection of the angle between a line connecting the midpoint of the acromion processes and the midpoint of the anterior-superior-iliac-spines (ASISs) and the global vertical axis. Additionally, Hunt *et al.* (2008) calculated the angle of lateral pelvic tilt separately from the trunk as the angle between a line drawn between the ASIS markers with respect to the global horizontal axis as seen in Figure 1.3. There are currently no accepted standards for tracking the motion of the trunk and pelvis.

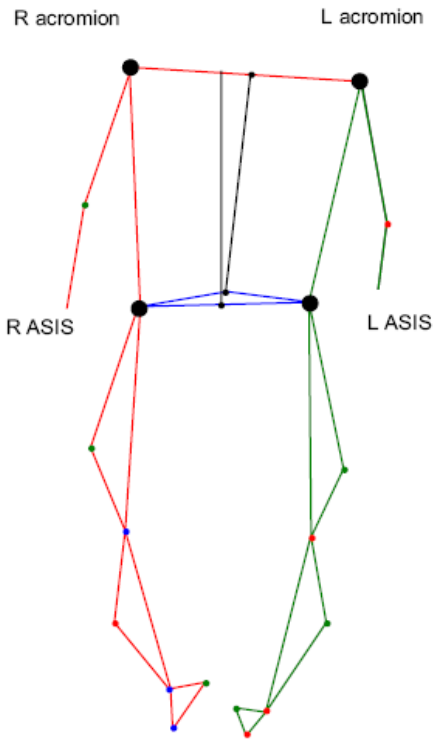

Figure 1.3: Trunk lean angle (Hunt *et al.*, 2008). Trunk lean was defined as the angle from vertical of a line connecting the midpoints of the acromion processes and the midpoints of the ASISs.

1.8. Sources and measurement of inter subject variability in gait data

Chau *et al.* (2005) defines variability in quantitative gait analysis as the fluctuation in the value of a kinematic, kinetic, spatio-temporal, or electromyographic measurement. Gait measurement variability has been classified as either extrinsic or intrinsic, the former being factors which can be controlled with quality improvements in the design, and the latter being factors which are naturally occurring and can only be measured and managed (Figure 1.4) (Schwartz *et al.*, 2004). Various statistical methods for measuring the variability between subjects have been used in the interpretation of waveform data. These include the upper and lower standard deviation curves (Kurz and Stergiou, 2003), bootstrap-derived prediction bands (Duhamel *et al.*, 2004; Lenhoff *et al.*, 1999),

correlation coefficients, coefficient of variation (Chau *et al.*, 2005; Winter, 1984), and principal component analysis (PCA) (Deluzio and Astephen, 2007; Chau *et al.*, 2005).

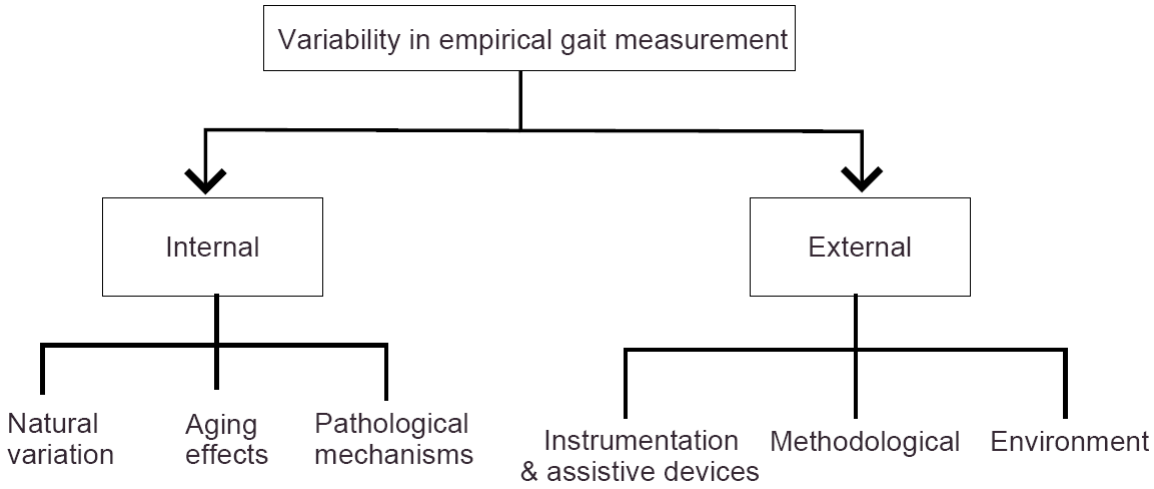

Figure 1.4: Sources of variability in empirical gait measurements (Chau *et al.*, 2005).

1.9. Principal Component Analysis

PCA is a multivariate statistical technique which has recently been shown to provide a novel approach to the analysis of kinematic and kinetic gait waveform parameters (Deluzio *et al.*, 1997; Deluzio and Astephen, 2007). PCA provides a unique analysis technique due to the ability to extract independent features of variation from groups of waveforms which can be used to statistically quantify differences between groups of subjects. PCA can be used to check for differences in waveform shape and magnitude between groups when typical discrete parameters such as ranges and peaks show no differences or prove difficult to pinpoint (Deluzio *et al.*, 1997).[2]

[2] Detailed PCA methodology can be found in Appendix C.

1.10. Purpose

Frontal plane thoracic tilt has recently come into consideration in the literature when discussing gait alterations in subjects with knee osteoarthritis as a possible mechanism to reduce the joint loading in the medial compartment of the knee. Although some studies have used motion analysis in order to quantify trunk lean, the literature lacks a comparison of the typical lateral bend gait pattern of subjects with and without knee osteoarthritis. This study aims to quantify and assess the contributions of both the thorax and the pelvis as separate segments to the gait alteration observed in subjects with knee osteoarthritis. Additionally lacking in the current literature is a comprehensive study providing information on associated factors such as knee and hip adduction moments in both control and osteoarthritis subjects.

An additional focus of this study is a method by which we can more accurately analyze joint angles in human motion data, given the individual uniqueness among subject posture. A new method to objectively reduce inter subject variability in joint angle data using PCA is presented using an example of thoracic tilt.

1.11. Objectives

The objectives of this work include:

1) To develop models which define frontal plane trunk lean and pelvic tilt during gait in order to:

   a. compare differences in thoracic and pelvic lateral tilt between osteoarthritis and control subjects

   b. compare the results of these models to those proposed previously in the literature

2) To compare the trunk and pelvic tilt data for the osteoarthritis and control groups by:

   a. analyzing four discrete mean parameters (20-80% stance, first peak, mid-stance, and second peak) for each definition

   b. applying PCA to the waveform data to detect differences in shape and magnitude

3) To objectively reduce the inter subject variability in the joint angle data caused by a difference in neutral angle between subjects in order to more accurately evaluate the differences in motion patterns between the osteoarthritis and control groups.

1.12. Hypotheses

1) Subjects with knee osteoarthritis lean their trunks farther over their stance limb during gait compared to control subjects.

2) Using PCA it is possible to objectively reduce the inter subject variability in joint angle waveform data caused by differences in neutral angles between subjects.

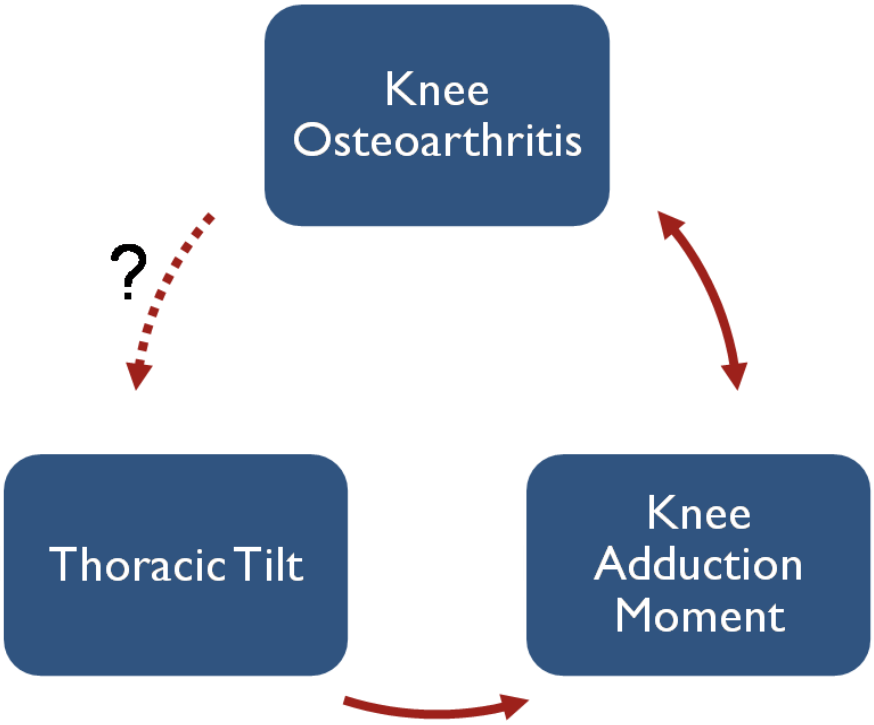

Figure 1.5: The connection of knee osteoarthritis, the knee adduction moment, and thoracic tilt. It is known that knee osteoarthritis is linked to increased knee adduction moments, and that by increasing thoracic tilt we can decrease the knee adduction moment. This leads us to question whether those with knee osteoarthritis display altered thoracic tilt patterns.

1.13. Thesis structure

Chapter 1 presents an introduction to the concepts presented throughout this thesis. It provides an overview of knee osteoarthritis and gait analysis, and the connection of knee osteoarthritis to the knee adduction moment, as well as gait alterations which have been identified in subjects with knee osteoarthritis. This chapter then discusses the hip adduction moment and its relationship with knee osteoarthritis and the knee adduction moment. The concept of lateral trunk lean is introduced along with its connection to the knee adduction moment, and differences in the literature in measuring lateral trunk and pelvic motion are reviewed. The currently proposed methods for measuring inter subject variability in gait data, as well as their sources are detailed and PCA is introduced. Finally, the purpose, objectives, and hypotheses of this thesis are presented.

Chapter 2 is a paper which was submitted to *Clinical Biomechanics* in June, 2009, titled "Differences exist in trunk lean patterns during gait in subjects with knee osteoarthritis compared to control subjects." The paper presents the work towards objectives 1 and 2, and addresses the first hypothesis. The thesis methodology and results pertaining to these objectives are presented along with a detailed discussion of the results. Although the paper was the collaborative work of a number of authors, the first author is primarily responsible for the hypotheses, analyses of the results, and writing of the paper.

Chapter 3 is a technical note in preparation for submission to *Clinical Biomechanics* titled "Removing subject specific bias in kinematic data using Principal Component Analysis: an example using Thoracic Tilt." The technical note presents the work towards objective 3, and addresses the second hypothesis. The thesis methodology and results pertaining to this objective are presented along with a detailed discussion of the results.

This chapter is presented as a new method for reducing the inter subject variability in waveform data. It is designed in the same manner as the paper before it following the structure of introduction, methods, results, and discussion.

Chapter 4 is the concluding chapter of this thesis. It summarizes the thesis work and presents the implications and limitations of the study, along with the recommended future work.

Additional information is provided in the appendices. This includes detailed methodology; gait analysis, laboratory setup, and additional thoracic tilt definitions, as well as an in depth description of PCA with extended results from all thoracic tilt models.

Chapter 2

2. Differences exist in trunk lean patterns during gait in subjects with knee osteoarthritis compared to control subjects

Heather S Linley

Elizabeth A Sled

Elsie G Culham

Kevin J Deluzio

Submitted to *Clinical Biomechanics* in June, 2009

2.1. Introduction

The adduction moment at the knee has received significant attention in the study of the pathomechanics of knee osteoarthritis (OA) (Baliunas *et al.*, 2002; Hurwitz *et al.*, 2002). It has been shown to be a good surrogate for evaluating medial compartment loading at the knee joint (Baliunas *et al.*, 2002; Hurwitz *et al.*, 2002; Hurwitz *et al.*, 1998) and is related to the severity and progression of OA (Astephen *et al.*, 2008; Miyazaki *et al.*, 2002).

Gait alterations have been linked with changes in joint loading in the lower limb (Mundermann *et al.*, 2005). By increasing trunk lean towards the stance limb by about 10 degrees, healthy subjects were able to reduce their first peak knee adduction moments an average of 65% (Mundermann *et al.*, 2008). Hunt *et al.* (2008) evaluated knee adduction moments and lateral trunk lean during the stance phase of gait in 120 people with medial knee OA. The authors found negative correlations between the knee adduction moment and lateral trunk lean angle at both the first and second adduction moment peaks, however the study lacked a control subject group for comparison. A recent study with a small sample size found no difference in trunk lean angles between those with unilateral or bilateral knee OA, or healthy subjects (Tanaka *et al.*, 2008). Lateral trunk lean has been found to be correlated with peak knee adduction moments, and an increase in trunk sway has been found to reduce the first peak knee adduction moment. There is a need to compare lateral trunk movement between control and OA groups in larger samples.

Pelvic control in the frontal plane has been attributed to the strength of the hip abductor muscles (MacKinnon and Winter, 1993). In an 18 month longitudinal study, Chang *et al.* (2005) evaluated internal hip abduction moments to quantify the magnitude of hip abductor muscle torque generation in 57 subjects with mild medial knee OA. Results of the study showed that a greater internal hip abduction moment at baseline lowered the odds of progression of knee OA. The authors speculated that weakness of the hip abductor muscles causes a drop in the pelvis on the swing side and shifts the centre of mass farther over the swing limb, thereby increasing medial compartment loading in the knee (Chang *et al.*, 2005). Mundermann *et al.* (2005) found that patients with severe knee OA had significantly lower first and second peak external hip adduction moments compared to the matched control subjects, while subjects with less severe knee OA had hip adduction moments throughout stance that were similar to their matched controls. The authors suggested that subjects with less severe knee OA have sufficient hip abductor muscle strength to maintain an altered position of the trunk laterally over the support limb, while those with more severe knee OA lack this muscle strength and drop the pelvis on the contralateral side. A drop of the pelvis would lead to a lean of the trunk away from the support limb, resulting in a higher first peak knee adduction moment. A study with 32 unilateral knee OA subjects compared knee and hip adduction moments between the affected and contralateral limbs (Briem and Snyder-Mackler, 2009). The authors found that the first peak hip adduction moment on the involved side was significantly smaller than that of the contralateral side, but found no significant interlimb differences in peak knee adduction moment. The authors suggested that an explanation for this low hip

adduction moment is a lateral trunk lean. However, neither motion of the pelvis and trunk, nor hip muscle strength was measured in these studies.

Although lateral trunk lean has been proposed by numerous authors as a potential gait alteration in subjects with knee OA as a mechanism to reduce knee joint loading, there lacks a connection between the contributing factors such as knee and hip adduction moments, and pelvic and thoracic motion. The purpose of this study is to test the hypotheses put forth by previous authors speculating on the effects of lateral trunk lean in subjects with knee OA in order to develop a better understanding of the underlying mechanics affecting this disease.

We hypothesized that subjects with knee OA alter their gait as a mechanism to reduce the joint loading at the knee by leaning their trunks farther over their stance limb compared to controls. This alteration is thought to be a combination of both pelvic and thoracic tilt. This study will contribute to the development of a further understanding of the clinical implications of lateral trunk lean in knee OA patients.

2.2. Methods

2.2.1. Participants

Participants were recruited locally through advertisements and by recommendation of local orthopaedic surgeons. Subjects included those with medial compartment knee OA and a control group of healthy subjects. The study was approved by the University Health

Sciences Research Ethics Board and each subject provided informed consent prior to participating. Inclusion criteria for the OA group were age (≥ 40 years), diagnosis of knee OA by a physician, subject-reported pain in the knee joint during most days of the month, and one of either radiographic evidence of medial compartment OA or evidence of medial compartment cartilage loss verified by arthroscopy or magnetic resonance imaging. Participants were excluded if the lateral knee compartment was more affected than the medial. If subjects were found to have bilateral OA, the most affected limb was used as the test limb. Participant exclusion criteria included intra-articular corticosteroid or visco-supplementation injection into either knee within a previous 3 month period, significant co-morbidities including heart disease, stroke, and active cancer treatment, OA caused by other arthritis types, a history of avascular necrosis, previous peri-articular fractures of the knee joint, Paget's disease, villonodular synovitis, joint infection, neuropathic arthropathy, acromegaly, Wilson's disease, hemocromatosis, gout or recurrent pseudogout, and osteopetrosis (Dieppe *et al.*, 1995).

Control subjects were healthy adults with a negative clinical diagnosis of knee OA, hip OA, rheumatoid arthritis, and no reports of knee or hip pain or knee trauma. All control subjects were screened radiographically to verify a negative diagnosis of knee OA. Subjects from the OA group were matched with subjects from the control group based on gender and age primarily, followed by an agreement between subject pairs in height. The test leg for each control subject was assigned to correspond with the test leg of the matched OA subject.

Radiographic OA disease severity was graded using the Kellgren and Lawrence (KL) scale while pain, stiffness, and physical function in the OA group were assessed using the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) for which the reliability and validity has been established for these populations (Bellamy *et al.*, 1988).

2.2.2. Gait Analysis[3]

Gait analysis was performed on all subjects. Two Optotrak motion capture cameras (Optotrak 3020, Northern Digital Inc., Waterloo, ON, Canada) were used to track participants as they walked along an 8 m walkway. Three dimensional kinematic data were recorded at a sample rate of 100 Hz. Marker clusters containing infrared light emitting diodes (IREDs) were placed on the dorsum of the foot over the metatarsals, lateral shank, lateral thigh, sacrum, and over the spinous processes of the 7^{th} cervical/1^{st} thoracic vertebrae. Two AMTI (Advanced Mechanical Technology Inc., Watertown, MA, USA) force platforms embedded in the walkway recorded the subject's ground reaction forces and moments at a sampling rate of 200 Hz. Participants provided and wore their own comfortable walking shoes during testing. Subjects were instructed to walk at a self-selected speed along the walkway. Five trials were obtained for each subject in which the foot of the test limb successfully landed on the force platform, and all markers were visible by the cameras.

[3] A detailed methodology of gait analysis is provided in Appendix A.

The locations of bony landmarks in relation to marker clusters were recorded using a digitizing pointed probe fitted with a cluster of markers. The landmarks were found by palpation performed by a physiotherapist. Landmarks included the first metatarsal heads, the fifth metatarsal heads, the medial and lateral malleoli, the medial and lateral epicondyles of the knee, the greater trochanters, points on the pelvis directly vertical to the greater trochanters at the level of the mid-iliac crest bilaterally, and the acromion processes of the scapulae.

An inverse dynamics approach (Visual 3D, C-Motion, Germantown, MD, USA) was used to calculate frontal plane moments for the knee and hip joints. Segment coordinate systems were defined for both the pelvis and the thorax. Joint moments and pelvic and thoracic tilt angles in the frontal plane were calculated across the stance phase of gait for both subject groups. Individual subject trials were event- and time-normalized to 100% of stance and then averaged at each percent of stance to provide one mean curve per subject. For each of the two subject groups, the mean subject curves were then ensemble averaged to provide a mean curve for each group across the stance phase of gait (MATLAB, The Math Works Inc., Natick, MA, USA).

2.2.3. Pelvic and Thoracic Tilt Definitions[4]

Figure 2.1 shows a visual representation of the definition used to describe pelvic tilt and two definitions of thoracic tilt. The angular motion of the pelvis was defined using

[4] Additional definitions of thoracic tilt were developed. These are described in Appendix B.

Cardan angles as previously suggested by Kadaba *et al.*, 1990. Lateral pelvic tilt was calculated as the frontal plane rotation of the pelvis coordinate system in reference to the lab coordinate system. The Cardan angle sequence of sagittal pelvic rotation – lateral pelvic tilt – transverse pelvic tilt was used as recommended by Baker (2001). The angle of thoracic tilt was calculated using a two dimensional planar projection similar to previous studies (Hunt *et al.*, 2008; Tanaka *et al.*, 2008). Thoracic tilt was defined in two ways (Figure 2.1 b, c): (i) the orientation of the thorax was measured with respect to the coordinate system of the lab (Thoracic Tilt – Lab) and also (ii) with respect to the coordinate system in the pelvis (Thoracic Tilt – Pelvis). The thorax was defined by a line connecting the midpoints between the pelvis landmarks and the acromion processes. The angle of thoracic tilt was first calculated by projecting the distal-proximal line up the thorax onto the frontal plane of the lab coordinate system and tracking it relative to the lab vertical. A second definition of thorax tilt used the same distal-proximal thorax line created from the landmarked points, but this time it was monitored throughout stance as a projection onto the frontal plane of the pelvis coordinate system. All angles and moments are reported during the stance phase of gait for the affected limb. Pelvic and thoracic tilt angles are reported as positive for a tilt over the stance limb.

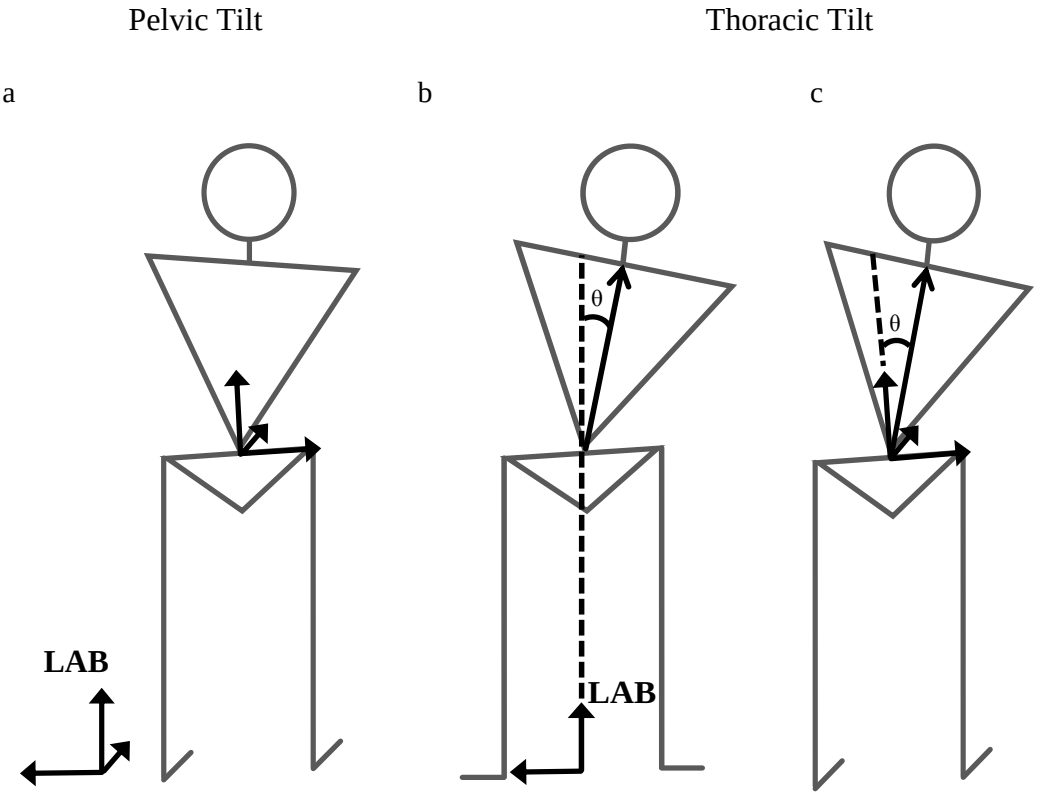

Figure 2.1: Pelvic and thoracic tilt definitions. (a) Pelvic tilt defined as the lateral motion of the pelvis coordinate system with respect to the lab coordinate system, (b) two dimensional planar rotation of the thorax as a projection onto the frontal plane of the lab coordinate system from vertical (Thoracic Tilt – Lab), (c) two dimensional planar rotation of the thorax as a projection onto the frontal plane of the pelvis coordinate system from vertical (Thoracic Tilt – Pelvis).

2.2.4. Statistical Analysis

Four discrete parameters were extracted from the gait measure waveforms:

1) Means were calculated across the mid portion of stance (20-80% stance)

2) Peak values were recorded for both groups in the first phase of stance

3) Peak values were recorded for both groups in the second phase of stance

4) Midstance values (50% stance) for both groups were compared

Student's t-tests were performed to detect differences between the OA and control groups ($\alpha = 0.05$).

In order to detect shape and magnitude differences in the gait data, PCA was performed on all waveforms (Deluzio *et al.*, 1997).[5] The first three principal components (PCs), representing the majority of the variation in the data were analyzed for significant differences between groups. They were also interpreted for physical meaning and statistical significance.

2.3. Results

Eighty subjects participated in the study, split evenly between the control and OA groups (23 females in each group). Thirty-three of the 40 OA subjects had bilateral medial compartment knee OA. Significant differences were found between the two groups in weight, body mass index (BMI), gait speed, cadence, and double limb support time, however no differences were found in age, height, or stride length ($p \leq 0.05$). A summary of these results is found in Table 1.

Radiographic KL scores for the OA subjects were as follows:

- 7 subjects: KL score of 4
- 9 subjects: KL score of 3
- 20 subjects: KL score of 2
- 4 subjects: KL score of 1

[5] Detailed PCA methodology can be found in Appendix C.

Radiographic KL scores for the control subjects were as follows:

- 2 subjects: KL score of 2
- 7 subjects: KL score of 1
- 31 subjects: KL score of 0

The WOMAC scores for pain, stiffness, and physical function are graded on scales from 0-20, 0-8, and 0-68 respectively. The OA group was found to have a pain score of 5.55 (2.87), a stiffness score of 3.08 (1.80), and a physical function score of 19.60 (11.77) reported as mean (standard deviation).

Table 1. Subject demographics and gait speed

	Group	N	Gender	Age	Height (m)	Weight (kg)*	BMI*	Gait Speed (m/s)*
Mean (SD)	Control	40	23 F	64 (9)	1.70 (0.09)	69.7 (11.0)	24.0 (3.2)	1.12 (0.19)
	OA	40	23 F	63 (10)	1.73 (0.11)	82.3 (20.0)	27.4 (5.5)	1.00 (0.2)
P-value				0.59	0.23	0.001	0.001	0.006

* Statistically significant difference between groups are noted for $\alpha = 0.05$.
BMI = Body Mass Index

Hip and knee adduction moments, as well as pelvic tilt and thoracic tilt – pelvis displayed double peak curves (Figure 2.2). However, no discernable second peak could be identified for thoracic tilt – lab.

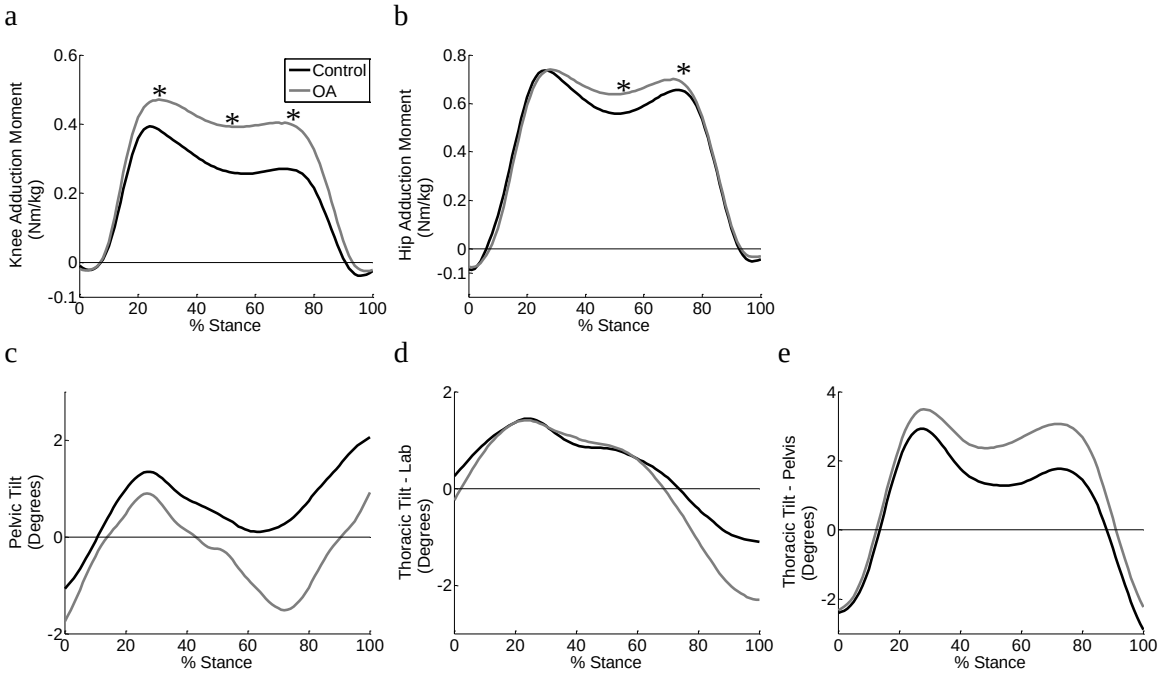

Figure 2.2: Group mean curves for OA and control groups across the stance phase of gait for the affected limb as indicated for (a) the external knee adduction moment (b) the external hip adduction moment (c) pelvic tilt (d) thoracic tilt - lab (e) thoracic tilt - pelvis. Positive values in (a) and (b) indicate adduction moments. Positive values (c), (d) and (e) indicate a tilt over the stance limb. Significant differences in discrete parameters (peaks and means) are indicated by * ($p < 0.05$).

Knee and hip adduction moments were found to be significantly higher for the OA group across the mid portion of stance (20-80% stance), at the midstance (50% stance) point, and in the second peak. The first peak knee adduction moment was also significantly higher for the OA group (Table 2). Most discrete measures of pelvic and thoracic tilt showed the OA group to have greater peak and mean values than the control group, however these differences were not significant.

Table 2. Discrete parameter values and comparisons between the two subject groups

	Means (SD)		p-value
Group	Control	Osteoarthritis	
Knee Adduction Moment (Nm/kg)			
20-80% Stance *	0.30 (0.10)	0.41 (0.14)	$p < 0.001$
First Peak *	0.41 (0.11)	0.52 (0.15)	$p < 0.001$
Midstance (50%) *	0.26 (0.10)	0.39 (0.15)	$p < 0.001$
Second Peak *	0.29 (0.11)	0.43 (0.15)	$p < 0.001$
Hip Adduction Moment (Nm/kg)			
20-80% Stance *	0.63 (0.08)	0.67 (0.10)	$p = 0.03$
First Peak	0.77 (0.11)	0.79 (0.12)	$p = 0.37$
Midstance (50%) *	0.56 (0.10)	0.64 (0.11)	$p = 0.002$
Second Peak *	0.67 (0.10)	0.72 (0.12)	$p = 0.02$
Pelvic Tilt (Deg)			
20-80% Stance	0.6 (3.9)	-0.3 (3.7)	$p = 0.26$
First Peak	2.6 (4.0)	1.9 (4.1)	$p = 0.46$
Midstance (50%)	0.5 (4.0)	-0.2 (3.9)	$p = 0.41$
Second Peak	-0.4 (3.8)	-1.9 (3.9)	$p = 0.09$
Thoracic Tilt – Lab (Deg)			
20-80% Stance	0.7 (1.7)	0.7 (2.0)	$p = 0.84$
First Peak	1.9 (1.6)	1.9 (2.0)	$p = 0.92$
Midstance (50%)	0.8 (1.8)	0.9 (2.2)	$p = 0.88$
Thoracic Tilt – Pelvis (Deg)			
20-80% Stance	1.8 (5.4)	2.8 (3.3)	$p = 0.32$
First Peak	3.2 (5.5)	4.1 (3.4)	$p = 0.42$
Midstance (50%)	1.3 (5.3)	2.4 (3.3)	$p = 0.29$
Second Peak	1.9 (5.3)	3.3 (3.5)	$p = 0.18$

* Statistically significant difference between groups are noted for $\alpha = 0.05$.

PCA detects shape and magnitude differences in the waveform data. Since each PC is independent of the others, we can examine them individually for differences between groups. In the model of pelvic tilt and both models of thoracic tilt over 97% of the total variation in the data was captured in the first three PCs.

The first PC captures the largest source of variation in the data. The loading vectors for PC1 have positive coefficients throughout the entire stance cycle for all models of pelvic and thoracic tilt. As seen in previous studies, this describes an overall magnitude feature in the data (Deluzio and Astephen, 2007). Therefore, PC1 captures a measure of the average angle throughout stance. This angle is a measure of an individual subject's average neutral posture. The magnitude variation in the data is related to measuring absolute segment angles, which vary from person to person. Therefore, the largest source of variation in the waveform data is due to differences in neutral posture between subjects and is unrelated to the actual motion pattern or shape of the waveforms. In all three models, there was no difference in PC1 between the control and OA groups. Due to the independence of PCs, PC1 provides a unique and objective means of removing the variation due to neutral position.

The second PC describes a difference in waveform shape and reveals statistically significant differences between the control and OA groups in pelvic tilt and both definitions of thoracic tilt (Figure 2.3). The loading vectors for all models display positive values in early stance and negative values in late stance. This captures a range of motion in pelvic and thoracic tilt. Comparing the subjects with high and low z-scores shows that a high PC2 score corresponds to an increased difference in motion between the early and late stages of stance. The OA subjects demonstrated an increased range of motion during stance for both pelvic tilt and thoracic tilt – lab ($p < 0.05$). The control subjects had an increased range of motion during stance for thoracic tilt – pelvis ($p < 0.05$).

The third PC also describes the waveform shape but does not show significant differences between the control and OA groups in any of the three models. PC3 captures the waveform's positive and negative peaks.

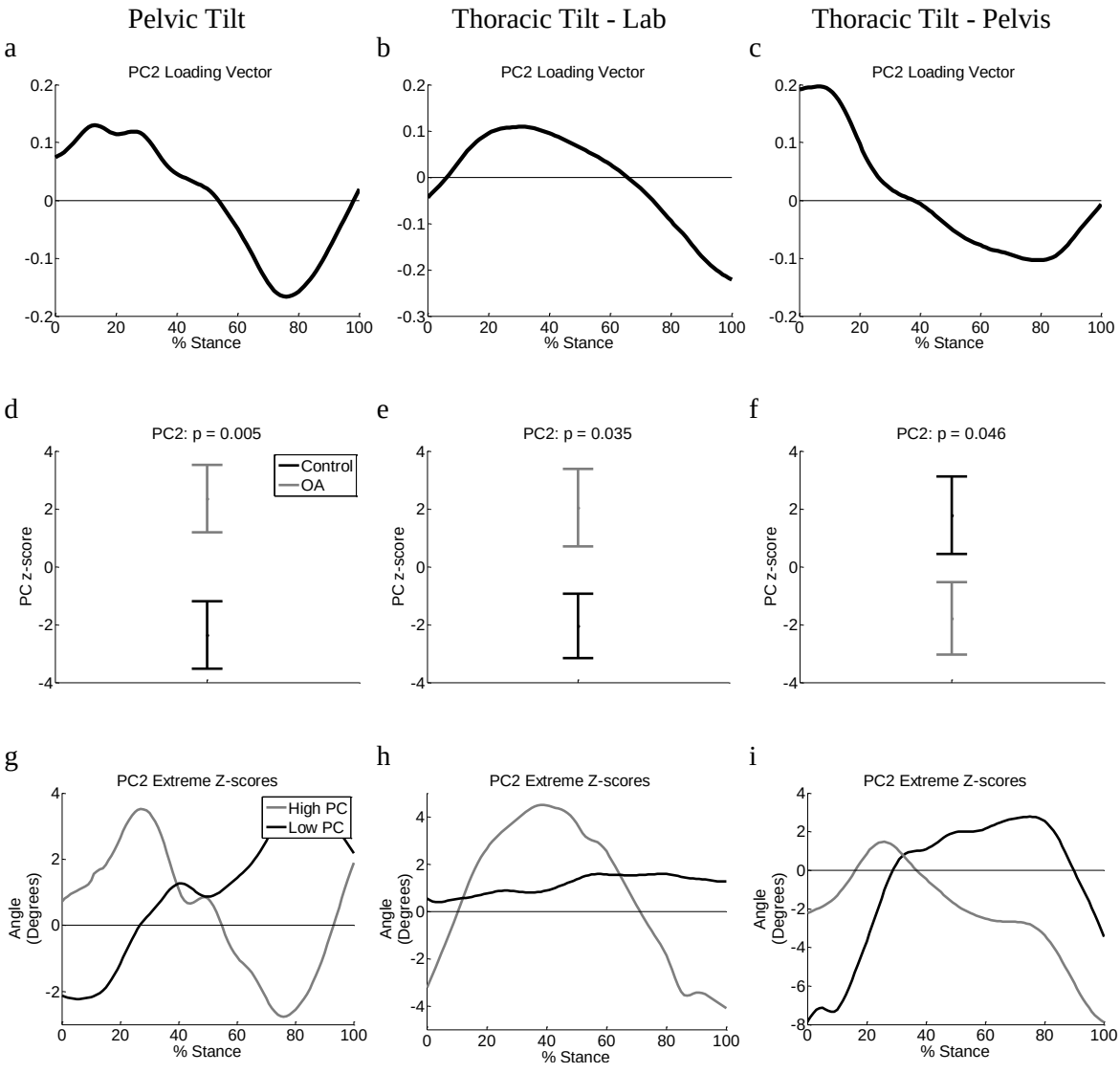

Figure 2.3: PCA loading vectors for (a) pelvic tilt, (b) thoracic tilt – lab, (c) thoracic tilt – pelvis. PCA interaction plots for (d) pelvic tilt, (e) thoracic tilt – lab, (f) thoracic tilt – pelvis. PCA high and low z-scores for (g) pelvic tilt, (h) thoracic tilt – lab, (i) thoracic tilt – pelvis.

Another way of illustrating the features captured by the second PC is by reconstructing the original waveforms. The visual representation of pelvic and thoracic tilt is based entirely on the reconstructed data having removed the bias due to the variation in neutral position between subjects (Figure 2.4). Figure 2.5 shows that subjects with OA began the stance phase with their pelvis raised slightly less on the stance side compared to control subjects. The pelvis then dropped on the stance side in both groups and the thorax tilted over the stance limb. At approximately 75% of the stance phase both groups dropped their pelvis on the swing side; however, this was significantly more pronounced in the OA group. Near the end of stance the pelvis was raised on the swing side for both groups, and the thorax was tilted over the swing side, but this thoracic tilt was much larger for the OA subjects.

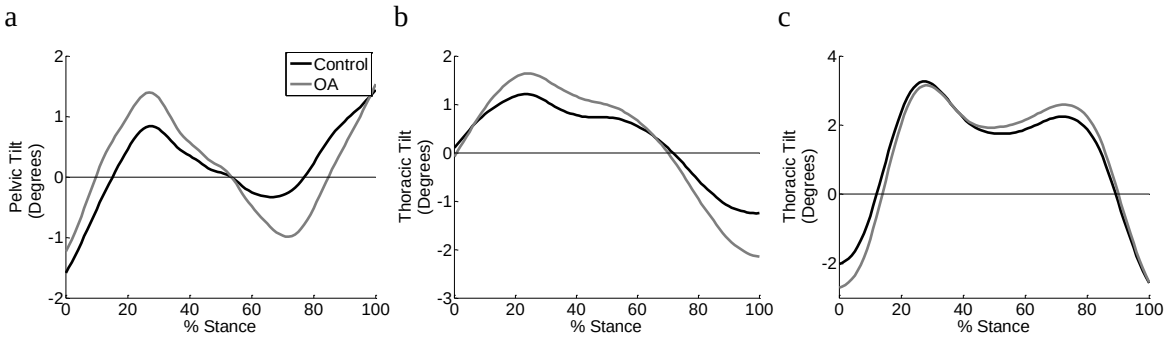

Figure 2.4: Reconstructed data using PC2 for (a) pelvic tilt, (b) thoracic tilt - lab, and (c) thoracic tilt - pelvis. A positive angle reflects a tilt over the stance limb.

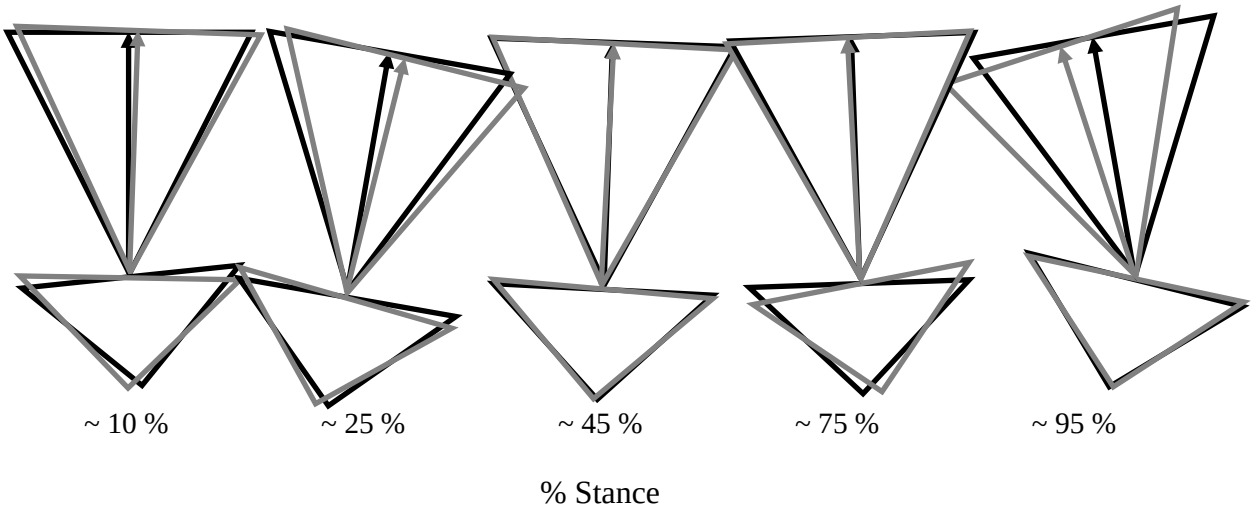

Figure 2.5: Visual represntation of the variation in motion of the thorax and the pelvis throughout the stance cycle in the control (black) and OA (grey) subjects. Each visualization represents the rear view with the stance limb on the right, and the swing limb on the left. Motions are exagerrated beyond actual angle values for effective visualisation.

2.4. Discussion

Knee and hip adduction moments were significantly higher in the OA group compared to controls, while discrete measures found no differences between groups in pelvic or thoracic tilt. However, PCA detected significant group differences in range of motion in pelvic and thoracic tilt. Therefore, differences do exist in pelvic and thoracic tilt between subjects with knee OA and control subjects.

The knee adduction moment of OA subjects was found to be higher than control subjects throughout the stance phase of gait as measured by all discrete parameters. These observations are consistent with previous studies. Mundermann *et al.* (2005) found that subjects with more severe OA displayed a significantly higher first peak knee adduction moment than both the control group and the less severe OA group. They also found that

the more severe OA group had significantly higher second peak knee adduction moments compared with the less severe OA group. The less severe knee OA group had a significantly lower second peak knee adduction moment compared to the control group ($p < 0.05$). Astephen *et al.* (2008) argued that the mid-stance knee adduction moment provides a better measure of the differences between control and OA gait patterns as it is a speed-independent measure. They found that both the moderate OA and severe OA group had significantly higher mid-stance knee adduction moments than the control group ($p < 0.002$). It has also been reported that the overall magnitude of the adduction moment captured by PC1 is not affected by speed differences between groups (Landry *et al.*, 2007).

The external hip adduction moment was found to be higher throughout stance phase in all discrete parameters except the first peak. This agrees with Huang *et al.* (2008) who found that subjects with severe OA displayed significantly higher mid-stance and second peak internal hip abduction moments compared to controls. Alternatively, two studies have reported lower external hip adduction moments in OA subject groups compared to controls (Astephen *et al.*, 2008; Mundermann *et al.*, 2005). Astephen *et al.* (2008) reported a significant difference in the first peak hip adduction moment between the control group and both a moderate and a severe OA group, with the controls having a higher peak than both OA groups ($p < 0.002$). Similarly, Mundermann *et al.* (2005) reported that severe OA subjects had significantly lower hip adduction moments in the first and second peaks of stance compared to the control groups. Interestingly, although their less severe OA subjects did show slightly higher hip adduction moments for both

peaks during stance compared to controls, the differences were not statistically significant. Some theories have been put forth to explain the results of these data including speculation about hip abductor muscle strength (Chang *et al.*, 2005; Mundermann *et al.*, 2005). However, none of the reviewed studies have collected hip abductor muscle strength data in order to justify these suggestions. Yamada *et al.*, (2001) found no difference in isometric hip abductor strength between OA and control groups.

The equivocal results of these studies and lack of explanatory data suggests that other factors may affect the hip adduction moment throughout stance such as gait speed, BMI, and disease severity. In fact, a recent study found that gait velocity and subject mass were the main contributors to the frontal plane hip moment, while hip abductor strength and gluteus medius muscle activation explained only a small amount of variability in the hip adduction moment waveform during midstance (Rutherford and Hubley-Kozey, 2009).

Definitions of pelvic tilt vary significantly between studies. It has been suggested that an optimal sequence of rotations for the pelvis is axial rotation, lateral tilt, and transverse tilt in order to correspond to conventionally defined anatomical terms (Baker, 2001). Baker demonstrated that measures of lateral pelvic tilt correlate better to relative hip height using the proposed sequence rather than the conventional transverse pelvic tilt, lateral pelvic tilt, sagittal pelvic rotation sequence. We chose this proposed sequence in order to best interpret our results clinically. Our lateral pelvic tilt data show a progressive increase in tilt over the stance limb reaching the first peak at approximately 25% of stance phase. The second peak shows a tilt over the swing side which occurs at about 75% of

stance. These results are similar to the findings of Hunt *et al.* (2008) who reported a peak rise of the pelvis on the swing side at 23.1% stance in 120 OA patients, and a peak drop of the pelvis on the swing side at 81.5% stance. Our lateral pelvic tilt data did not compare well in waveform shape to studies by Huang *et al.* (2008), Crosbie *et al.* (1997) or Whittle *et al.* (1999) which all used different sequences of rotation to define lateral pelvic tilt. However, it is known that the selected sequence of rotations highly influences the values of the Cardan angles (Baker, 2001).

A standard definition for tracking the lateral motion in the thorax has not been proposed. Some studies have tracked the thorax at two positions along the spine relative to each other and the pelvis (Crosbie *et al.*, 1997b). Others have defined a virtual line up the centre of the body using the midpoints between landmarked points on the pelvis and shoulders and tracked this line relative to the vertical (Hunt *et al.*, 2008). Tanaka *et al.* (2008) monitored the lateral bend in the thorax by tracking a line between C7 and S1 of the spine relative to a line perpendicular to the floor. Our OA group data for thoracic tilt – lab correspond well to that reported by both Hunt et al (2008) and Crosbie *et al.* (1997b). Whittle *et al.* (1999) defined lateral bend similarly to our model of thoracic tilt – pelvis, however, they recorded the motion of the lower portion of the thorax at the thoracolumbar junction of the spine. Our thoracic tilt angles and our pattern of motion match well with the model of Whittle *et al.* (1999).

The combined motions of the pelvis and thorax showed both segments having more motion compared to the lab coordinate system in the OA group; however, when the

thorax was projected on the pelvis coordinate system, the control subjects showed a significantly greater range of motion throughout stance phase. This indicates that the OA subjects moved the trunk and pelvis more as a single unit, whereas the control group balanced motions between the two segments. Although it has been suggested that due to the small range of motion in the pelvis, lateral thoracic tilt is primarily achieved by leaning the trunk and shoulders (Hunt *et al.*, 2008), the thorax displayed an equally small range of lateral tilt motion as the pelvis. These results lead us to believe that lateral trunk lean is a combination of both pelvic and thoracic tilt. Even a small angular tilt could move the mass of the trunk sufficiently to alter the loading in the knee, which is consistent with suggestions by Chang *et al.* (2005) that a drop in the pelvis on the swing side could shift the body's mass towards the swing limb, increasing forces across the medial compartment of the knee in the stance limb.

When comparing pelvic motion between groups, questions arise as to whether the motion pattern differences are being caused exclusively by a significant difference in walking speed between the OA and control subjects. Subjects were asked to walk at a self-selected speed in order to obtain the best representation of their true gait patterns. Control subjects were found to have a significantly higher walking speed than OA subjects. One way to deal with this issue is to remove this speed effect statistically using an analysis of covariance (ANCOVA). However, the appropriateness of this method has been questioned for a study of this nature because speed, which acts as the covariate in the model, may affect OA severity and progression, which acts as the treatment in the model, violating the assumptions of ANCOVA (Astephen *et al.*, 2008). Additionally,

frontal plane pelvic tilt has been shown to increase significantly with an increase in gait speed (Crosbie *et al.*, 1997a). We found that the OA group, which ambulated at a slower gait speed, displayed increased lateral tilt in the pelvis throughout stance compared to the control group. This suggests that if the two groups had equivalent gait speed, the difference between the OA and control groups would have been larger. Therefore the difference in pelvic motion found between the two subject groups is not due to gait speed differences.

The radiographic data indicated that these subjects represent a moderate OA group with an average KL score of 2.5 out of 4 and individual subject scores varied widely within that range (scores of 1-4). Subjects with differences in disease severity have been shown to display gait pattern differences (Astephen *et al.*, 2008). These results may not be representative of the patterns found in subjects with very mild or very severe knee OA. Although two control subjects were found to have KL scores of 2 and seven subjects were found to have scores of 1, these subjects did not have any knee pain or limitations, and had not suffered any previous knee injury. Thus, they had a negative clinical diagnosis of knee OA and were included in our control set.

In summary, patients with OA in the medial compartment of the knee displayed greater knee and hip adduction moments as well as altered gait patterns in both the thorax and the pelvis throughout the stance cycle of gait compared to controls. Patients with OA tended to display an increased range of motion in their trunk and pelvis, and tended to move these two segments more as a single unit compared to control subjects. In

recognizing our limitations to conclusively discuss causality in our data, we recommend future studies be designed to test the effects of controlled pelvic and thoracic tilt on hip and knee adduction moments in subjects with knee OA.

Acknowledgements

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Chapter 3

3. Removing subject specific bias in kinematic data using Principal Component Analysis: an example using Thoracic Tilt

Heather S Linley

Kevin J Deluzio

Prepared for submission to *Clinical Biomechanics* 2009

3.1. Introduction

Human motion data presents challenges for researchers and clinicians when trying to recognize patterns and differences in groups of subjects. Each individual is unique and as such introduces variability into the data. The variability between subjects in human motion data offers a complex challenge. Without removing this variability, biomechanical differences may be difficult to detect. One must take caution when reducing the inter subject variability to ensure that the method is objective and does not impose further bias. Measuring absolute joint angles in human motion data poses a unique problem as it requires the identification of a reference orientation such as a neutral posture from which to base the measurements (O'Dwyer *et al.*, 2009). Often, a neutral position can be more easily defined for joints of the lower limb than for the trunk and pelvis when measuring joint angles because the hip, knee and ankle angles are all relative angles, whereas the trunk and pelvis angles are absolute angles (Davis *et al.*, 1991). In order to compare kinematic data between subjects, the bias measured in a neutral posture must be removed for each subject. Specific methods for reducing variability due to subject differences have been previously suggested. Several authors have proposed the removal of a common static angular offset for each subject (Al-Eisa *et al.*, 2006a; Al-Eisa *et al.*, 2006b; Davis *et al.*, 1991; Kadaba *et al.*, 1990). Subjects are asked to stand erect while motion capture data is collected in order to define a standard neutral posture angle, which is then removed before comparative analysis. Additionally, researchers often remove human motion trials from the dataset which are considered to be outliers, even if the pattern of motion is found to be consistent with other trials (Chau *et al.*, 2005). One such study

demonstrated that by combining curve registration and outlier removal, the variability among a group of knee angle curves as estimated by both standard deviation and bootstrapping was minimized (Chau *et al.*, 2005). However, it should be noted that the knee angle curve which was discarded by the authors displayed the same pattern of motion as the other curves. Although this method succeeds in reducing the variability between trials, relevant and useful data is lost in the process.

PCA is a statistical tool used to extract independent features from waveform data corresponding to shape and magnitude (Deluzio *et al.*, 1997). PCA linearly transforms the original data variables into new, mutually orthogonal representations of those vectors. Each subject is associated with a principal component (PC) score based on a projection of that subject's data onto each particular loading vector. Subject groups can then be tested for significant differences in individual PCs. PCA allows us to establish which independent features represent the most variability in the data as well as providing a means of testing for differences between groups beyond traditional discrete parameters (Ramsay and Silverman, 1997).

We propose a new method, based on PCA, for reducing the inter subject variability in human motion data. We demonstrate how this can be useful in detecting subject group differences.

3.2. Methods

Gait analysis was performed on 80 subjects, 40 of whom were verified clinically and radiographically to have medial compartment knee osteoarthritis (OA). The other 40 subjects were the control group and were age and gender matched with the OA subjects. The angle of thoracic tilt was calculated for each subject with respect to the global laboratory based reference frame using a two dimensional planar projection. The thorax was defined by a line connecting the midpoints between two landmarks on the pelvis and the acromion processes. The angle of thoracic tilt was calculated by projecting the distal-proximal line up the thorax onto the frontal plane of the laboratory coordinate system. The data were normalized across the stance cycle of gait where the affected limb was in stance. A positive angle reflects a tilt over the stance limb.

PCA was applied to the thoracic tilt waveform data in order to detect magnitude and shape differences between the control and OA subject groups across the stance cycle of gait. This is achieved by transforming the original variables into new, mutually orthogonal representations of these variables. The first three PCs were examined as they represented majority of the variability in the data. Each PC was analyzed for significant differences between the two subject groups.

The waveform data were reconstructed using the second and third PCs. This was achieved by summing the multiplication of the loading vectors of each PC by each individual subject PC-score. The reconstruction provided a physical representation of the original waveform data with the inter subject variability objectively reduced.

The coefficient of variation was calculated for both the control and OA subject groups for the original unaltered data and for the reconstructed data using Eq. (1) (Winter, 1984):

$$CV = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N}\sigma_i^2}}{\frac{1}{N}\sum_{i=1}^{N}\left|\overline{X}_i\right|} \times 100\% \qquad (1)$$

where CV is the coefficient of variance, N is the number of intervals over the stance phase, $\overline{X}_i$ is the mean waveform at the *i*th interval, and σ_i is the standard deviation about $\overline{X}_i$. Finally, the differences between the coefficients of variation in the raw data and the reconstructed data were calculated.

3.3. Results

Figure 3.1 shows the raw data for both the control and OA subjects individually, as well as the group means. Subjects display similar waveform shapes, but the waveforms vary in magnitude. This suggests a variation between subjects in the average thoracic tilt angle which is related to the neutral position.

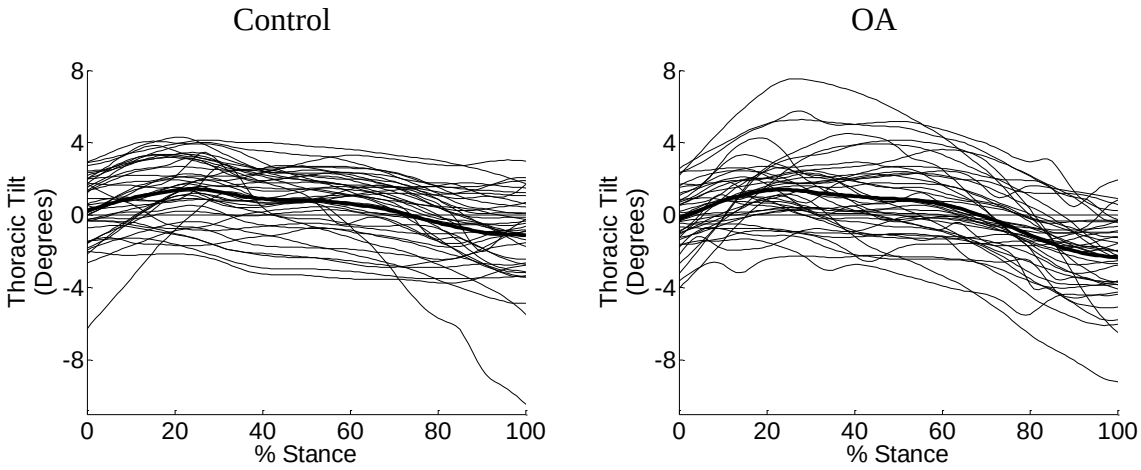

Figure 3.1: Thoracic tilt curves for all control and all OA subjects, and group means for control and OA subjects across the stance phase of gait.

The coefficients of variation for the raw data are as follows:

$CV_{Control}$ = 230.6%

CV_{OA} = 194.5%

The results from the first three PCs can be found in Figure 3.2. The loading vector for PC1 remains positive throughout the stance cycle. Furthermore, the values of the loading vector are of near equal magnitude. This indicates that it represents a magnitude effect in the data (Deluzio and Astephen, 2007). A subject with a high PC1 score has a high magnitude waveform, while a subject with a low PC1 score has a low magnitude waveform. This captures the variation due to the average joint angle throughout stance which is related to individual differences in neutral posture. PC1 represents the greatest source of variation in the data. Due to the independence of PCs, the other PCs can be analyzed independent of this variability. We can remove this overall magnitude effect in the data by excluding the first PC from our analysis. PCA provides an objective method for removing the inter subject variability in the data by identifying magnitude differences between subjects in the first PC.

The second PC represents a difference operator with positive coefficients throughout the first half of the stance phase of gait, and negative coefficients in late stance. It describes the range of motion in the data. Subjects with high PC2 scores display a substantial range of motion between early and late stance. This PC is the only one of the examined three which is found to be different between the groups. It has found a significant difference in the range of motion between the control and the OA groups.

The third PC also describes a range of motion. A subject with a high PC3 score displays a peak in the mid portion of the stance cycle. This PC was not found to be statistically significant between the subject groups.

The first three PCs represented over 98% of the variability in the waveform data. PC1, PC2 and PC3 captured 73.9%, 16.5%, and 7.8% of the variation in the waveform data respectively.

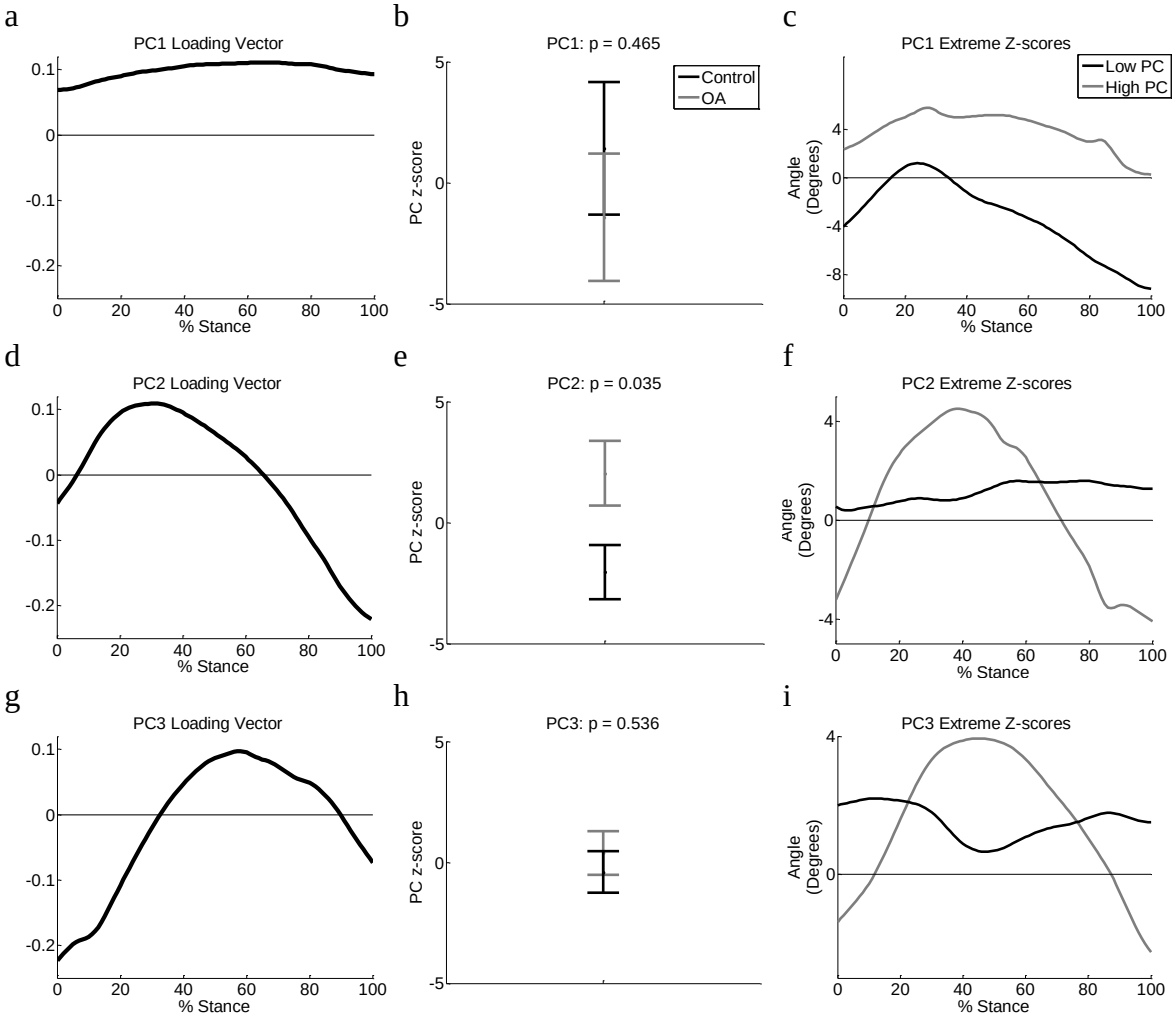

Figure 3.2: PCA loading vectors, PCA interaction plots, and PCA extreme Z-scores for thoracic tilt data. Loading vectors for (a) PC1, (d) PC2, and (g) PC3. Interaction plots for (b) PC1, (e) PC2, and (h) PC3. Extreme Z-scores for (c) PC1, (f) PC2, and (i) PC3. PC1 accounts for 73.9% of the overall variation in the data while PC2 and PC3 account for 16.4% and 7.8% respectively.

The reconstructed data using PC2 and PC3 provides a physical representation of the significant differences between the two subject groups. Figure 3.3 shows the reduced inter subject variability in the data set after excluding the first PC. This reconstruction allows us to interpret the physical motion of the subjects relative to one another.

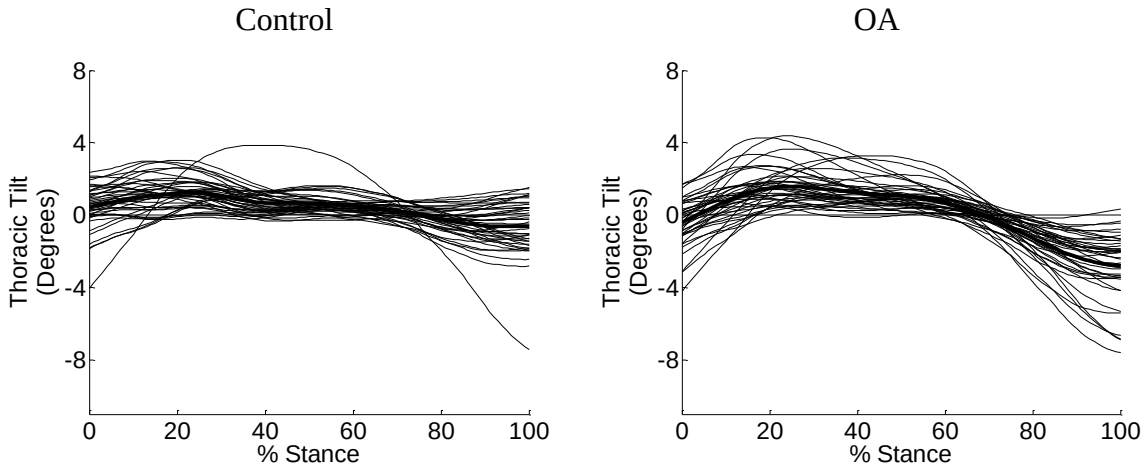

Figure 3.3: Reconstructed thoracic tilt data using PC2 and PC3. [6]

The coefficients of variation for the reconstructed data are as follows:

$CV_{Control}$ = 131.3%

CV_{OA} = 84.3%

The overall reduction in variation between the raw data and the reconstructed data is 99.3% for the control group, and 110.2% for the OA group.

3.4. Discussion

Conventional methods for reducing inter subject variability in human motion data include removing outliers, or defining and removing a common offset. Often outliers are defined as any waveform which is found to be more than two standard deviations away from the mean (Chau *et al.*, 2005). This however poses problems in the study of human motion data, particularly in the study of angle data. Absolute angles are difficult to

[6] The control data set still contains one possible outlier. This subject displays a difference in the pattern of motion of their gait compared to the other subjects.

measure and joint angles tend to be subject specific and depend on the individual's neutral posture and position.

There is the potential for a bias to be introduced when removing a common static offset based on a calibration trial. When using a static calibration trial to normalize subject specific angle data, the definition of a neutral position may be arbitrary. This method of removing inter subject variability can introduce bias and be subjective. Even a small change in a subject's neutral posture could be discriminatory. Our technique using PCA provides the ability to test for this effect.

Implications can arise from analyzing human motion data without accounting for variability between the subjects. Variability adds noise to the data and makes it difficult to detect differences between groups. Reporting unaltered human motion data can also have clinical implications. Using PCA we developed an objective method for removing the inter subject variability due to joint angle bias. Although this technique is demonstrated for thoracic tilt, the method is valid for use with any joint angle calculation where the primary source of variability in the data is a difference in magnitude between subjects.

Chapter 4

4. Summary and Conclusions

4.1. Summary

This thesis aimed to contribute towards the development of a greater understanding of the underlying mechanics of knee osteoarthritis, and the gait alterations adopted naturally by subjects with pathologies. It has been suggested that subjects with knee OA lean their trunks farther over their stance limb laterally during gait compared to control subjects as a mechanism to reduce knee joint loading (Chang *et al.*, 2005; Hunt *et al.*, 2008; Mundermann *et al.*, 2005; Mundermann *et al.*, 2008). The current literature lacks a comparative study of patients and controls to verify this gait alteration. It was found in this study, that subjects with knee osteoarthritis do have a different pattern of motion than control subjects in both the trunk and the pelvis.

The secondary aim of this thesis was to present a new method of reducing variability in joint angle waveform data attributed to differences in neutral posture between subjects. It was shown for an example with thoracic tilt that using PCA, the inter subject variability can be objectively reduced in order to compare the patterns of motion of two subject groups.

Knee and hip adduction moments were calculated for all subjects, and the OA group was found to have higher moments of force in almost all of the four investigated discrete waveform parameters, except for the first peak hip adduction moment. This led to the

conclusion that subjects with OA do in fact have higher knee adduction moments than control subjects; however the hip adduction moment data in this study did deviate from some reports in the literature. A total of four models were developed to describe lateral thoracic tilt. Two of these models are described in Chapter 2, and two are found in Appendix B. One model was used to describe pelvic tilt. The thorax was defined both as a planar projection of a distal-proximal line up the centre of the body onto the frontal plane of both the lab and pelvis coordinate systems, and using a rigid body coordinate system to track the frontal plane rotation in reference to the lab and pelvis coordinate systems. The pelvis was defined using a rigid body coordinate system to track its frontal plane rotation in reference to the lab coordinate system. Differences between the two groups were analyzed for four discrete waveform parameters (peaks and mean). In all cases no significant differences were found between the control and OA groups. However, PCA found differences between the groups in the range of motion of the pelvis and thorax during gait. OA subjects were found to move the thorax and pelvis more as a single unit, while the control group balanced motions between the two segments. Therefore the first hypothesis could not be explicitly accepted, as it was found that the range and patterns of motion are more descriptive of the difference between groups than any discrete value.

Using PCA an objective method was developed to remove the effect of differences in neutral postures throughout the gait cycle in joint angle waveform data. This magnitude effect is captured entirely in the first PC as it represents the majority of the variation in the data. Therefore, by performing a reconstruction on the data excluding this PC, it is

possible to objectively remove the inter subject variability which confirms the second hypothesis.

4.2. Implications

Speculation on the possibility of trunk lean as a gait alteration in subjects with knee osteoarthritis has been put forth by numerous authors (Hunt *et al.*, 2008; Mundermann *et al.*, 2005; Mundermann *et al.*, 2008; Tanaka *et al.*, 2008; Briem and Snyder-Mackler, 2009). This is, however, the first study to quantify lateral trunk lean in both control and OA subjects in order to analyze differences between the subject groups. Additionally, there has been speculation regarding the hip abductor muscles and their effects on the motion of the pelvis (Chang *et al.*, 2005; MacKinnon and Winter 1993). However until now, the lateral motion of the pelvis had not been well documented with the Cardan angle sequence of rotation proposed by Baker in 2001.

The newly proposed method of reducing inter subject variability caused by a difference in neutral posture has the potential to alter the way that we approach subjects in joint angle data who appear to be outliers, or data which has a large amount of variation. This new approach provides the ability to objectively reduce this variability, without losing any subjects, in order to more accurately analyze the clinical significance of the results. Until now, PCA has been used as a method of reducing variability by ignoring the low variance PCs. For example, if over 97% of the variation in the waveform data was captured in the first four PCs, then the remaining PCs would be discarded. This

study is the first which uses PCA to remove the non discriminatory variation or unwanted variability which is the largest source of variation in the waveform data (PC 1).

4.3. Limitations

It is unknown from the results of this study whether lateral trunk lean as a gait alteration is related causatively to the pathology, or whether this gait alteration is adopted as a mechanism to improve quality of living such as reducing joint pain, or increasing gait speed. Although we can speculate on the quantifiable benefits associated with an increased lateral sway (Mundermann *et al.*, 2008), we cannot determine from this study the reason for this change in gait pattern in OA subjects.

It is also not possible to know the long term effects of this gait alteration found in subjects with OA. It is equally as possible that the difference in motion in subjects with OA contributes to the progression of the pathology, as it is possible that this alteration slows the progression of the disease.

A challenge with the results for pelvic tilt in the present study was the variation in the literature in the selected sequence of rotations for the representation of the Cardan angles. The variation in rotation sequence has a large influence on the results of the comparison between the two groups, and can change the pattern of motion during gait. Therefore it is extremely important to identify the sequence of rotations which will allow the most accurate clinical interpretation of the results.

Inherent limitations are associated with performing human motion analysis. Marker position is greatly affected by skin motion. We attempt to account for this by securing marker clusters over large muscle masses where skin motion is minimized. However, this motion cannot be eliminated. Additionally, problems develop using marker clusters attached to fins which are secured to the body using tape or belts. When attempting to measure very small angles in the trunk and pelvis, a small deviation of the fin marker clusters could have a large effect on the reported angles. It makes sense then, by this limitation, that we focus more on the differences in the pattern of motion between the control and OA groups, rather than the angular values associated with each model. Errors are also associated with the identification of bony landmarks on the body. These digitized points are found by palpation which should be done by an experienced clinician. However, landmarks are often difficult to locate, particularly on subjects with a higher BMI. As knee OA is known to be associated with a higher BMI, this influences the current study. The incorrect identification of landmarks can alter the determination of joint centres and segment coordinate systems.

4.4. Conclusions

Subjects with OA in the medial compartment of the knee have greater knee and hip adduction moments compared with control subjects. They also display an altered pattern of motion in the thorax and pelvis throughout the stance cycle of gait. Patients with OA tend to have an increased range of motion in the two segments; however the thorax and

pelvis tend to move more as a single unit in the OA subjects whereas the control subjects balance this motion between the segments.

Using PCA we developed a new method for objectively reducing the inter subject variability in joint angle data caused by a difference in neutral posture between subjects. This allows a more accurate clinical analysis of the results based on the pattern of motion of the subject groups.

4.5. Future research and recommendations

A longitudinal study of subjects with knee OA to evaluate the biomechanical changes in the knee and hip adduction moments as well as thoracic and pelvic tilt would lead to a better understanding of the implications of the gait alteration. Other factors which should be measured longitudinally are clinical and radiographic evidence of OA progression, gait speed and BMI changes, and lower limb muscle strength changes. In designing a longitudinal study, a separate study should be performed during any gait analysis to quantify intra tester repeatability. This is important when performing landmarking by palpations on many subjects, especially those subjects who may not have easily identifiable bony landmarks.

In order to discuss causality between knee and hip adduction moments and thoracic and pelvic tilt, a designed experiment to test the effects of quantifiably controlled pelvic and thoracic tilt on knee and hip adduction moments should be performed. Similarly,

lateral trunk lean variation among distinct populations should be identified. For example, differences may exist between those with mild, moderate, and severe OA, or between men and women. This may influence the treatment of each of these groups.

Designing a comparative study to this one which involves the use of surface EMG to measure muscle contributions during gait would contribute to understanding the reasons behind the increase in range of motion of the trunk segments. Specifically, the motion of the pelvis is thought to be controlled primarily through the hip abductors (MacKinnon and Winter, 1993). A strengthening or weakening of these muscles in subjects with OA could then have an influence on their patterns of motion.

Many different models of thoracic tilt have been presented in this thesis. An evaluation of these models in order to develop a standard for measuring this joint angle would allow better comparisons in the literature.

Finally, it will be important to test the proposed method to reduce inter subject variability on other joint angles as well as other kinematic and kinetic waveform data. This will allow us to further verify the validity of the method, and will help to convince other researchers of its usefulness.

Chapter 1 References

Andrews, M., Noyes, F. R., Hewett, T. E., Andriacchi, T. P., (1996). Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: a critical analysis of the reliability of gait analysis data. J.Orthop.Res. 14, 289-295.

Astephen, J. L., Deluzio, K. J., (2009). Techniques in Modern Gait Analysis and Their Application to the Study of Knee Osteoarthritis. In: Leondes, C. T. (Ed.), Muscular Skeletal Systems. World Scientific Publishing Co. Pte. Ltd., pp. 39-72.

Astephen, J. L., Deluzio, K. J., Caldwell, G. E., Dunbar, M. J., (2008). Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity. J.Orthop.Res. 26, 332-341.

Baliunas, A. J., Hurwitz, D. E., Ryals, A. B., Karrar, A., Case, J. P., Block, J. A., Andriacchi, T. P., (2002). Increased knee joint loads during walking are present in subjects with knee osteoarthritis.

Brandt, K. D., (1998). The importance of nonpharmacologic approaches in management of osteoarthritis. Am.J.Med. 105, 39S-44S.

Buckwalter, J. A., Stanish, W. D., Rosier, R. N., Schenck, R. C., Jr., Dennis, D. A., Coutts, R. D., (2001). The increasing need for nonoperative treatment of patients with osteoarthritis.
Clin.Orthop.Relat Res. 36-45.

Chang, A., Hayes, K., Dunlop, D., Song, J., Hurwitz, D., Cahue, S., Sharma, L., (2005). Hip abduction moment and protection against medial tibiofemoral osteoarthritis progression. Arthritis Rheum. 52, 3515-3519.

Chang, A., Hurwitz, D., Dunlop, D., Song, J., Cahue, S., Hayes, K., Sharma, L., (2007). The relationship between toe-out angle during gait and progression of medial tibiofemoral osteoarthritis. Ann.Rheum.Dis. 66, 1271-1275.

Chau, T., Young, S., Redekop, S., (2005). Managing variability in the summary and comparison of gait data. J.Neuroeng.Rehabil. 2, 22.

Deluzio, K. J., Astephen, J. L., (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: an application of principal component analysis. Gait.Posture. 25, 86-93.

Deluzio, K. J., Wyss U.P., Zee B., Costigan P.A., Sorbie C., (1997). Principal component models of knee kinematics and kinetics: Normal vs pathological gait patterns. pp. 201-2.

Duhamel, A., Bourriez, J. L., Devos, P., Krystkowiak, P., Destee, A., Derambure, P., Defebvre, L., (2004). Statistical tools for clinical gait analysis. Gait Posture 20, 204-212.

Felson, D. T., Lawrence, R. C., Dieppe, P. A., Hirsch, R., Helmick, C. G., Jordan, J. M., Kington, R. S., Lane, N. E., Nevitt, M. C., Zhang, Y., Sowers, M., McAlindon, T., Spector, T. D., Poole, A. R., Yanovski, S. Z., Ateshian, G., Sharma, L., Buckwalter, J. A., Brandt, K. D., Fries, J. F., (2000). Osteoarthritis: new insights. Part 1: the disease and its risk factors. Ann.Intern.Med. 133, 635-646.

Hunt, M. A., Birmingham, T. B., Bryant, D., Jones, I., Giffin, J. R., Jenkyn, T. R., Vandervoort, A. A., (2008). Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis. Osteoarthritis.Cartilage. 16, 591-599.

Hurwitz, D. E., Ryals, A. B., Case, J. P., Block, J. A., Andriacchi, T. P., (2002). The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. J.Orthop.Res. 20, 101-107.

Hurwitz, D. E., Sumner, D. R., Andriacchi, T. P., Sugar, D. A., (1998). Dynamic knee loads during gait predict proximal tibial bone distribution. J.Biomech. 31, 423-430.

Kurz, M. J., Stergiou, N., (2003). The spanning set indicates that variability during the stance period of running is affected by footwear. Gait Posture 17, 132-135.

Lenhoff, M. W., Santner, T. J., Otis, J. C., Peterson, M. G., Williams, B. J., Backus, S. I., (1999). Bootstrap prediction and confidence bands: a superior statistical method for analysis of gait data. Gait Posture 9, 10-17.

Mundermann, A., Asay, J. L., Mundermann, L., Andriacchi, T. P., (2008). Implications of increased medio-lateral trunk sway for ambulatory mechanics. J.Biomech. 41, 165-170.

Mundermann, A., Dyrby, C. O., Andriacchi, T. P., (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. Arthritis Rheum. 52, 2835-2844.

Mundermann, A., Dyrby, C. O., Hurwitz, D. E., Sharma, L., Andriacchi, T. P., (2004). Potential strategies to reduce medial compartment loading in patients with knee osteoarthritis of varying severity: reduced walking speed. Arthritis Rheum. 50, 1172-1178.

Rutherford, D. J., Hubley-Kozey, C., (2009). Explaining the hip adduction moment variability during gait: Implications for hip abductor strengthening. Clin.Biomech.(Bristol., Avon.) 24, 267-273.

Sasaki, T., Yasuda, K., (1987). Clinical evaluation of the treatment of osteoarthritic knees using a newly designed wedged insole. Clin.Orthop.Relat Res. 181-187.

Schache, A. G., Blanch, P., Rath, D., Wrigley, T., Bennell, K., (2002). Three-dimensional angular kinematics of the lumbar spine and pelvis during running. Hum.Mov Sci. 21, 273-293.

Schwartz, M. H., Trost, J. P., Wervey, R. A., (2004). Measurement and management of errors in quantitative gait data. Gait Posture 20, 196-203.

Sharma, L., Hurwitz, D. E., Thonar, E. J., Sum, J. A., Lenz, M. E., Dunlop, D. D., Schnitzer, T. J., Kirwan-Mellis, G., Andriacchi, T. P., (1998). Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. Arthritis Rheum. 41, 1233-1240.

Tanaka, K., Miyashita, K., Urabe, Y., Ijiri, T., Takemoto, Y., Ishii, Y., Ochi, M., (2008). Characteristics of trunk lean motion during walking in patients with symptomatic knee osteoarthritis. Knee. 15, 134-138.

Taylor, N. F., Goldie, P. A., Evans, O. M., (1999). Angular movements of the pelvis and lumbar spine during self-selected and slow walking speeds. Gait.Posture. 9, 88-94.

Teichtahl, A. J., Morris, M. E., Wluka, A. E., Baker, R., Wolfe, R., Davis, S. R., Cicuttini, F. M., (2006). Foot rotation--a potential target to modify the knee adduction moment. J.Sci.Med.Sport 9, 67-71.

Whittle, M., Levine, D., (1999). Three-dimensional relationships between the movements of the pelvis and lumbar spine during normal gait. pp. 681-692.

Winter, D. A., (1984). Kinematic and Kinetic Patterns in Human Gait: Variability and Compensating Effects. Hum.Mov Sci. 3, 51-76.

Yasuda, K., Sasaki, T., (1987). The mechanics of treatment of the osteoarthritic knee with a wedged insole. Clin.Orthop.Relat Res. 162-172.

Chapter 2 References

Astephen, J. L., Deluzio, K. J., Caldwell, G. E., Dunbar, M. J., (2008). Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity. J.Orthop.Res. 26**,** 332-341.

Baker, R., (2001). Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait Posture 13**,** 1-6.

Baliunas, A. J., Hurwitz, D. E., Ryals, A. B., Karrar, A., Case, J. P., Block, J. A., Andriacchi, T. P., (2002). Increased knee joint loads during walking are present in subjects with knee osteoarthritis. Osteoarthritis.Cartilage. 10**,** 573-579.

Bellamy, N., Buchanan, W. W., Goldsmith, C. H., Campbell, J., Stitt, L. W., (1988). Validation study of WOMAC: a health status instrument for measuring clinically important patient relevant outcomes to antirheumatic drug therapy in patients with osteoarthritis of the hip or knee. J.Rheumatol. 15**,** 1833-1840.

Briem, K., Snyder-Mackler, L., (2009). Proximal gait adaptations in medial knee OA. J.Orthop.Res. 27**,** 78-83.

Chang, A., Hayes, K., Dunlop, D., Song, J., Hurwitz, D., Cahue, S., Sharma, L., (2005). Hip abduction moment and protection against medial tibiofemoral osteoarthritis progression. Arthritis Rheum. 52**,** 3515-3519.

Crosbie, J., Vachalathiti, R., Smith, R., (1997a). Age, gender and speed effects on spinal kinematics during walking. Gait and Posture 13-20.

Crosbie, J., Vachalathiti, R., Smith, R., (1997b). Patterns of spinal motion during walking. pp. 6-12.

Deluzio, K. J., Astephen, J. L., (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: an application of principal component analysis. Gait.Posture. 25**,** 86-93.

Deluzio, K. J., Wyss U.P., Zee B., Costigan P.A., Sorbie C., (1997). Principal component models of knee kinematics and kinetics: Normal vs pathological gait patterns. pp. 201-2.

Dieppe, P. A., Buckwalter, J. A., Hascall V., Lohmander L.S., (1995). Standardization of methods used to assess the progression of osteoarthritis of the hip or knee joints. Osteoarthritic Disorders 481-496.

Hunt, M. A., Birmingham, T. B., Bryant, D., Jones, I., Giffin, J. R., Jenkyn, T. R., Vandervoort, A. A., (2008). Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis. Osteoarthritis.Cartilage. 16**,** 591-599.

Hurwitz, D. E., Ryals, A. B., Case, J. P., Block, J. A., Andriacchi, T. P., (2002). The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. J.Orthop.Res. 20**,** 101-107.

Hurwitz, D. E., Sumner, D. R., Andriacchi, T. P., Sugar, D. A., (1998). Dynamic knee loads during gait predict proximal tibial bone distribution. J.Biomech. 31**,** 423-430.

Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., (1990). Measurement of lower extremity kinematics during level walking. J.Orthop.Res. 8**,** 383-392.

Landry, S. C., McKean, K. A., Hubley-Kozey, C. L., Stanish, W. D., Deluzio, K. J., (2007). Knee biomechanics of moderate OA patients measured during gait at a self-selected and fast walking speed. J.Biomech. 40**,** 1754-1761.

MacKinnon, C. D., Winter, D. A., (1993). Control of whole body balance in the frontal plane during human walking. J.Biomech. 26**,** 633-644.

Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., Shimada, S., (2002). Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. Ann.Rheum.Dis. 61**,** 617-622.

Mundermann, A., Asay, J. L., Mundermann, L., Andriacchi, T. P., (2008). Implications of increased medio-lateral trunk sway for ambulatory mechanics. J.Biomech. 41**,** 165-170.

Mundermann, A., Dyrby, C. O., Andriacchi, T. P., (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. Arthritis Rheum. 52**,** 2835-2844.

Rutherford, D. J., Hubley-Kozey, C., (2009). Explaining the hip adduction moment variability during gait: Implications for hip abductor strengthening. Clin.Biomech.(Bristol., Avon.) 24**,** 267-273.

Tanaka, K., Miyashita, K., Urabe, Y., Ijiri, T., Takemoto, Y., Ishii, Y., Ochi, M., (2008). Characteristics of trunk lean motion during walking in patients with symptomatic knee osteoarthritis. Knee. 15**,** 134-138.

Chapter 3 References

Al-Eisa, E., Egan, D., Deluzio, K., Wassersug, R., (2006a). Effects of pelvic asymmetry and low back pain on trunk kinematics during sitting: a comparison with standing. Spine 31**,** E135-E143.

Al-Eisa, E., Egan, D., Deluzio, K., Wassersug, R., (2006b). Effects of pelvic skeletal asymmetry on trunk movement: three-dimensional analysis in healthy individuals versus patients with mechanical low back pain. Spine 31**,** E71-E79.

Chau, T., Young, S., Redekop, S., (2005). Managing variability in the summary and comparison of gait data. J.Neuroeng.Rehabil. 2**,** 22.

Davis, R., Ounpuu, S., Tyburski, D., Gage, J., (1991). A Gait Analysis Data Collection and Reduction Technique. Hum.Mov Sci. 575-587.

Deluzio, K. J., Astephen, J. L., (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: an application of principal component analysis. Gait.Posture. 25**,** 86-93.

Deluzio, K. J., Wyss U.P., Zee B., Costigan P.A., Sorbie C., (1997). Principal component models of knee kinematics and kinetics: Normal vs pathological gait patterns. pp. 201-2.

Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., (1990). Measurement of lower extremity kinematics during level walking. J.Orthop.Res. 8**,** 383-392.

O'Dwyer, N., Smith, R., Halaki, M., Rattanaprasert, U., (2009). Independent assessment of pattern and offset variability of time series waveforms. Gait Posture 29**,** 285-289.

Ramsay, J. O., Silverman, B. W., (1997). Functional data analysis. Springer, New York.

Winter, D. A., (1984). Kinematic and Kinetic Patterns in Human Gait: Variability and Compensating Effects. Hum.Mov Sci. 3**,** 51-76.

Chapter 4 References

Baker, R., (2001). Pelvic angles: a mathematically rigorous definition which is consistent with a conventional clinical understanding of the terms. Gait Posture 13, 1-6.

Briem, K., Snyder-Mackler, L., (2009). Proximal gait adaptations in medial knee OA. J.Orthop.Res. 27, 78-83.

Chang, A., Hayes, K., Dunlop, D., Song, J., Hurwitz, D., Cahue, S., Sharma, L., (2005). Hip abduction moment and protection against medial tibiofemoral osteoarthritis progression. Arthritis Rheum. 52, 3515-3519.

Hunt, M. A., Birmingham, T. B., Bryant, D., Jones, I., Giffin, J. R., Jenkyn, T. R., Vandervoort, A. A., (2008). Lateral trunk lean explains variation in dynamic knee joint load in patients with medial compartment knee osteoarthritis. Osteoarthritis.Cartilage. 16, 591-599.

MacKinnon, C. D., Winter, D. A., (1993). Control of whole body balance in the frontal plane during human walking. J.Biomech. 26, 633-644.

Mundermann, A., Asay, J. L., Mundermann, L., Andriacchi, T. P., (2008). Implications of increased medio-lateral trunk sway for ambulatory mechanics. J.Biomech. 41, 165-170.

Mundermann, A., Dyrby, C. O., Andriacchi, T. P., (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. Arthritis Rheum. 52, 2835-2844.

Tanaka, K., Miyashita, K., Urabe, Y., Ijiri, T., Takemoto, Y., Ishii, Y., Ochi, M., (2008). Characteristics of trunk lean motion during walking in patients with symptomatic knee osteoarthritis. Knee. 15, 134-138.

Appendix References

Deluzio, K. J., Astephen, J. L., (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis: an application of principal component analysis. Gait.Posture. 25, 86-93.

Kadaba, M. P., Ramakrishnan, H. K., Wootten, M. E., (1990). Measurement of lower extremity kinematics during level walking. J.Orthop.Res. 8, 383-392.

Appendix A

A. Gait Analysis

All gait analysis took place in a common laboratory using the same equipment for all subjects. Marker clusters containing infrared emitting diodes (IREDs) were placed in a particular configuration on each body, as shown in Figure A.1. As described in Chapter 2, body landmarks were digitized after palpation. Laboratory equipment included a walkway embedded with two force platforms, and two Optotrak motion capture cameras (Figure A.2). The cameras track the locations of the IREDs throughout the analysis.

Figure A.1: Marker cluster set-up and configuration. Blue dots represent individual IRED markers.

Joint Kinematics

Joint angles are defined as the relative motion between two rigid body segments. In this thesis, joint angles are calculated using two distinct methods: (i) the relative motion between two coordinate systems, and (ii) a two dimensional planar projection of one segment onto a single plane of another segment coordinate system. Each rigid body coordinate system is defined using three anatomical axes created from the position and orientation of surface markers and virtual landmarked points during gait analysis.

Three rigid bodies are of importance in this particular study; the thorax, the pelvis, and the laboratory. Each has a defined coordinate system. The laboratory coordinate system is defined by a vertical axis, perpendicular to the horizontal floor, an axis in the direction of motion along the walkway (anterior-posterior), and the cross product of these two axes in the medial-lateral direction.

The coordinate system in the thorax was defined using the left and right acromion processes as the primary medial-lateral axis, and a plane created from the acromion processes and two points on the pelvis to define a proximal-distal perpendicular axis. The cross product of these two axes defined the third axis in the thorax coordinate system.

Finally, the coordinate system in the pelvis was defined using two landmarked points on the pelvis to define the primary medial-lateral axis, and a plane created from the landmarked pelvis points and the two greater trochanters to define a proximal distal

perpendicular axis. The cross product of these two axes defined the third axis in the pelvis coordinate system.

As the subject moves throughout the stance cycle of gait, the coordinate systems move relative to one another. This transformation is described using Cardan angles (Kadaba *et al.*, 1990). These angles are found using the vectors on each segment which have known orientations, and the rotations about three axes corresponding to a specified sequence. The transformation can be described by the equation:

$$p_2 = Rp_1 \tag{2}$$

where p_1 is the vector defined on the reference segment, and p_2 is the vector defined on the moving segment. The rotation matrix, R, is used to resolve the joint angles β, α, γ:

$$[R] = \begin{bmatrix} r_{11} & r_{12} & r_{13} \\ r_{21} & r_{22} & r_{23} \\ r_{31} & r_{32} & r_{33} \end{bmatrix} \text{ where } \begin{aligned} \beta &= \sin^{-1}(r_{31}) \\ \alpha &= -\sin^{-1}\left(\frac{r_{32}}{\cos\beta}\right) \\ \gamma &= -\sin^{-1}\left(\frac{r_{21}}{\cos\beta}\right) \end{aligned} \tag{3}$$

Joint Kinetics

The three directional forces and three directional moments obtained from the force plate data along with the kinematic data obtained from the marker system allow us to use the inverse dynamics approach to calculate the forces acting on the joints of the lower limbs. Each segment is modelled as a rigid body with a body fixed local coordinate system located at the segmental centre of mass. To perform the analysis a number of

variables are required. These include segment masses, locations of the centre of mass of each segment, number of distal segments in the desired chain, and the number, location, and value of all applied external forces on the segments. Many of the inertial properties are estimated from existing anthropometric data based on the mass and height of the subject. Inverse dynamic analysis begins at the most distal segment, and moves proximally. It can be broken down into three distinct steps:

1) The kinematic position data obtained from the marker system is differentiated twice in order to obtain the velocity and acceleration values for each rigid body segment.

2) The resultant joint force at the proximal end of the segment of interest is determined using the law of motion $\sum F = \sum ma$, where *F* is force, *m* is mass, and *a* is acceleration. As an example, the resultant force at the ankle joint, $\vec{F}_{ankle}$, is calculated based on the foot segment:

$$\vec{F}_{ankle} = m_{foot}\vec{g} + m_{foot}\vec{a}_{foot} + \vec{F}_{GRF} \qquad (4)$$

where m_{foot} is the mass of the foot, $\vec{g}$ is the force of gravity, $\vec{a}_{foot}$ is the acceleration of the centre of mass of the foot, and $\vec{F}_{GRF}$ is the ground reaction force measured from the force platform.

3) The resultant moment of force at the proximal end of the segment of interest is determined using the rotational law of motion $\sum M = \sum Fd$, where *M* is the moment of force, *F* is the force, and *d* is the moment arm.

Again using the example of the ankle joint, the resultant moment, $\vec{M}_{ankle}$, is calculated using properties from the foot segment:

$$\vec{M}_{ankle} = \left(\vec{F}_{ankle} d_{ankle}\right) + \left(\vec{F}_{GRF} d_{GRF}\right) + \left(I_{foot} \vec{\alpha}_{foot}\right) + \vec{M}_{foot} \quad (5)$$

where d_{ankle} and d_{GRF} are the moment arm lengths of force at the ankle and point of ground reaction force, I_{foot} is the mass moment of inertia of the foot, $\vec{\alpha}_{foot}$ is the angular acceleration of the foot, and $\vec{M}_{foot}$ is the ground reaction moment measured from the force platform.

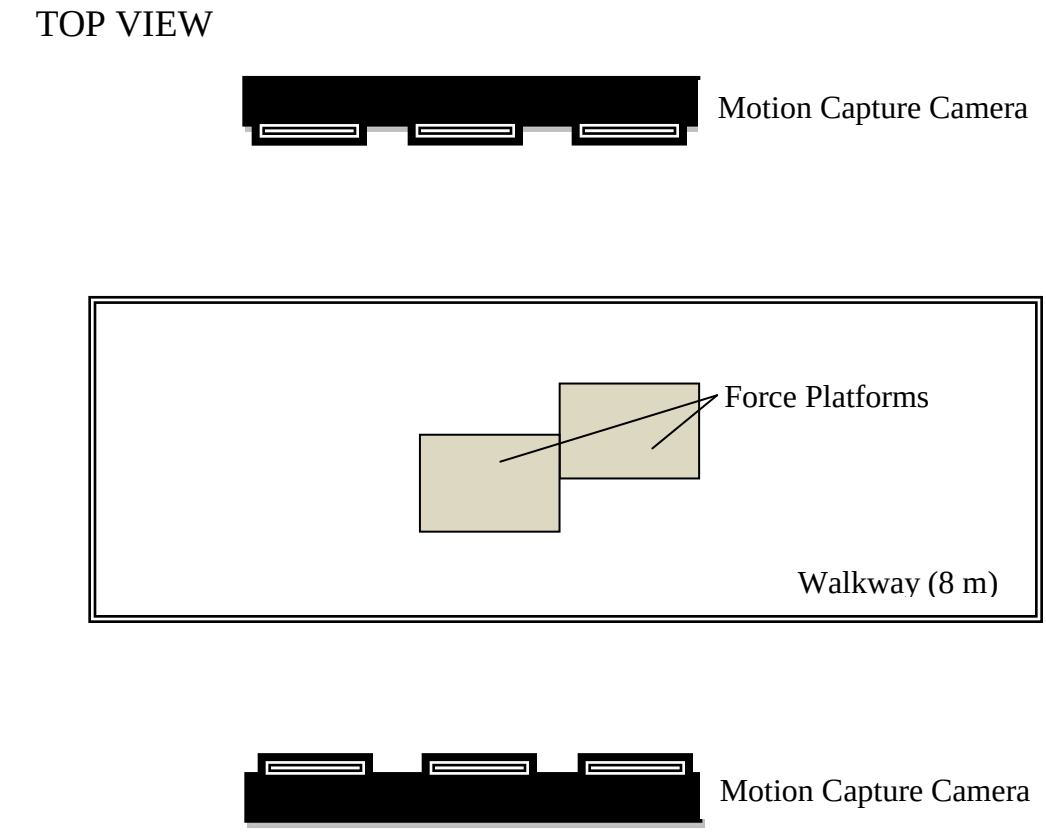

Figure A.2: Sketch of laboratory set-up (top view and side view). Note: Not to scale.

Appendix B

B. Other Thoracic Tilt Definitions

Two additional models of thoracic tilt were developed and analyzed which are not outlined in the main text (Figure B.1). These two definitions use the coordinate system in the thorax to calculate the lateral trunk lean angle with respect to: (i) the laboratory coordinate system and (ii) the pelvis coordinate system. Thoracic tilt was then calculated as the frontal plane rotation of the thorax coordinate system in reference to the laboratory coordinate system and the pelvis coordinate system using the Cardan angle sequence of lateral thoracic tilt – transverse thoracic tilt – sagittal thoracic rotation. The positive direction of lateral tilt is defined as being over the stance limb.

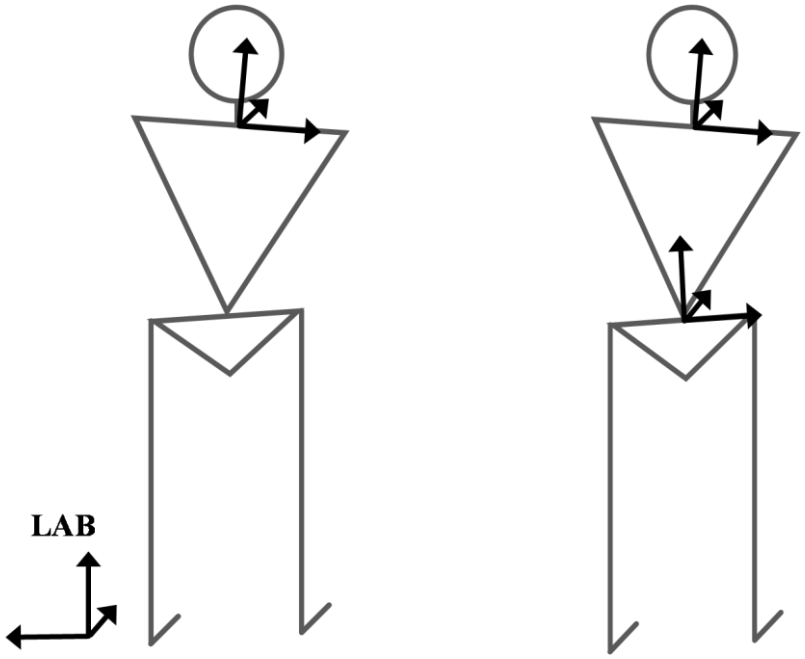

Figure B.1: Additional thoracic tilt definitions. On the left thoracic tilt is defined as the lateral motion of the thorax coordinate system with respect to the lab coordinate system. On the right, thoracic tilt is defined as the lateral motion of the thorax coordinate system with respect to the pelvis coordinate system.

Four discrete parameters were extracted from the thoracic tilt waveforms as outlined in Chapter 2. These values correspond to those often reported in the literature and include waveform peaks and mean values. Student's t-tests were performed to detect differences between the OA and control groups ($\alpha = 0.05$) and PCA was performed on the waveforms to detect shape and magnitude differences in the data. The first three PCs were extracted and interpreted for physical meaning and significance.

Figure B.2 shows the raw data waveform mean curves for the OA and control groups for the definition of thoracic tilt with respect to the lab coordinate system, as well as the second PC reconstruction of this data. Similarly, Figure B.3 shows the raw data waveform mean curves for the OA and control groups for the definition of thoracic tilt with respect to the pelvis coordinate system, as well as the PC2 reconstruction of the data. No significant differences were found between groups in any of the tested discrete parameters (Table 3).

Thoracic Tilt with respect to Lab

Raw: Reconstruction:

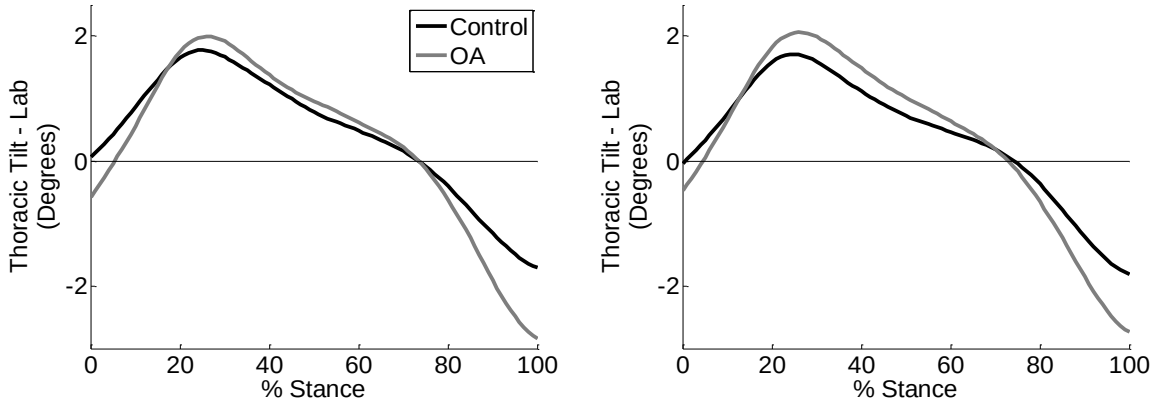

Figure B.2: Group mean curves for OA and control groups across the stance phase of gait for the affected limb as indicated for Thoracic Tilt - Lab (not projection). The raw data is presented in the figure on the left, and the PC2 reconstructed data is presented in the figure on the right.

Thoracic Tilt with respect to Pelvis

Raw: Reconstruction:

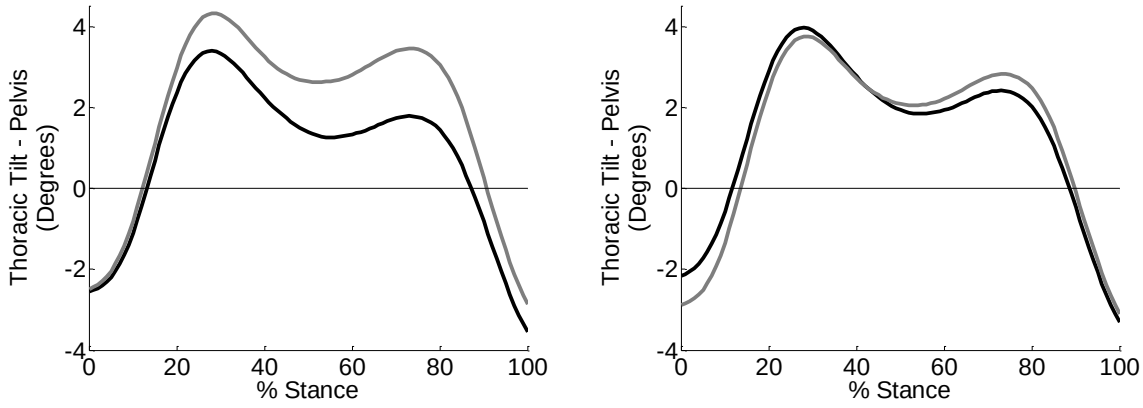

Figure B.3: Group mean curves for OA and control groups across the stance phase of gait for the affected limb as indicated for Thoracic Tilt - Pelvis (not projection). The raw data is presented in the figure on the left, and the PC2 reconstructed data is presented in the figure on the right.

Table 3. Discrete parameter values and comparisons between the two subject groups.

	Means (SD)		p-value
Group	Control	Osteoarthritis	
Thoracic Tilt – Lab (Deg)			
20-80% Stance	0.84 (2.41)	0.96 (2.82)	p = 0.83
First Peak	2.01 (2.26)	2.26 (2.77)	p = 0.65
Midstance (50%)	0.79 (2.49)	0.95 (3.11)	p = 0.79
Thoracic Tilt – Pelvis (Deg)			
20-80% Stance	2.04 (5.46)	3.29 (4.76)	p = 0.18
First Peak	3.73 (5.66)	4.90 (4.93)	p = 0.29
Midstance (50%)	1.39 (5.44)	2.64 (4.87)	p = 0.15
Second Peak	2.05 (5.39)	3.72 (4.92)	p = 0.10

Appendix C

C. Principal Component Analysis in detail

There are problems associated with reporting on subject group waveform differences based on discrete measures such as peaks and means. Waveform shape and magnitude varies between subjects and not all waveforms display definitive peaks. PCA provides a method with which to analyze the waveforms as a whole, capturing differences in shape and magnitude in the data. It is a powerful statistical tool which can be used for data reduction and to describe variability in the data. One of the most beneficial uses of PCA is the ability to extract the features in the data which explain the majority of the variability as independent components. Additionally, often the majority of the variation is captured within the first few PCs, easing the need for extended analysis.

As an example, the current study contained 80 subjects (40 OA, 40 control), each with 101 gait waveform points, each representing 1% of the stance phase of gait. This provides the structure of the data matrix:

$$\begin{bmatrix} & p & 1 & 2 & 3 & \dots & 101 \\ n & & & & & & \\ 1 & & x_{1,1} & x_{1,2} & x_{1,3} & & x_{1,101} \\ 2 & & x_{2,1} & x_{2,2} & x_{2,3} & & x_{2,101} \\ 3 & & x_{3,1} & x_{3,2} & x_{3,3} & & x_{3,101} \\ \vdots & & \vdots & \vdots & \vdots & & \vdots \\ 80 & & x_{80,1} & x_{80,2} & x_{80,3} & & x_{80,101} \end{bmatrix}$$

Here, *n* is the number of subjects, and *p* is the number of data points during the stance cycle of gait. The data is then transformed into a covariance matrix $[S]$ which removes the mean of the observations from each data point.

The next step is to extract the eigenvectors, or PC loading vectors from this covariance matrix. This is accomplished according to the equation:

$$(S - \lambda I)U = 0 \quad (6)$$

where λ is the vector of eigenvalues, *I* is the identity matrix, and *U* are the PCs. The percent variation explained by each PC is then calculated by dividing the eigenvalue for each PC by the sum of eigenvalues for all PCs.

It is often useful to know how different a particular subject's waveform is from the mean waveform. This observation, known as the subject's Z-score is used in comparative statistical testing. The Z-scores are calculated for each subject by multiplying the mean removed original data by each separate loading vector. This gives one subject Z-score for each independent PC. It is often beneficial to examine those subjects with extreme Z-scores for a particular PC in order to successfully interpret the extracted feature. For each model of thoracic tilt as well as for pelvic tilt, Z-scores between the OA and control groups were analyzed for statistical differences using Student's t-tests.

A procedure for reducing the inter subject variability in human motion data, particular in angle data, was developed using PCA. Since PCs are independent of one another, if it is found that a PC captures the overall magnitude difference in the data, we can use PCA to objectively reduce this variation by removing this PC. A PC describing a magnitude effect in the data can be identified by examining the loading vector. If the loading vector remains positive throughout the time cycle, and if the values remain near equal, the PC is determined to be an overall magnitude effect. This magnitude effect has been found previously to capture the most variation in the data for both joint angle and moment data (Deluzio and Astephen, 2007). In other words, the inter subject variability is often found to be the first PC, representing the greatest source of variability in the data.

The independence of PCs leads to a unique means of dealing with this variability. In order to remove the PC representing the inter subject variability; the data is reconstructed based on the PCs representing the next most variation. This is accomplished by summing the multiplication of the Z-scores of each subject by the loading vectors for each of the desired PCs:

$$R = \sum_{i=j}^{n} \left(Z_j U_j \right) \tag{7}$$

where R is the constructed data, n is the number of subject Z-scores, j are the desired PCs for inclusion in the reconstruction, and U are the loading vectors of each j.

The reconstructed waveforms can then be interpreted for physical meaning with the inter subject variability objectively reduced.

All additional PCA which was performed on the data and scrutinized for physical meaning can be found on the following pages. These data were used to verify and validate the method for reducing inter subject variability as described in Chapter 3, however their results are superfluous to the discussion of this thesis.

Pelvic Tilt (as defined in Chapter 2): Extended PCA

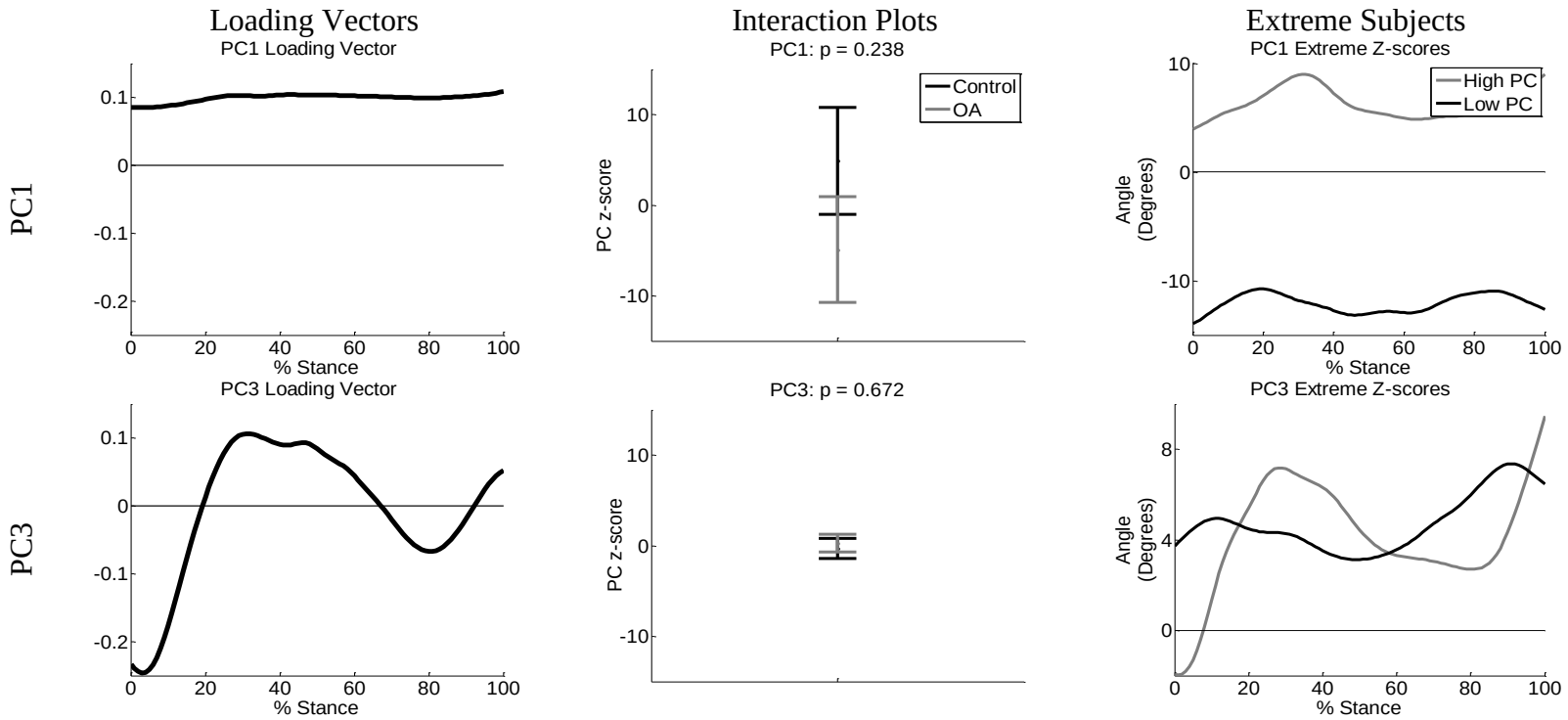

Figure C.1 . PCA loading vectors, PCA interaction plots, and PCA extreme Z-scores for PC1 (top row) and PC3 (bottom row) for pelvic tilt as defined in Chapter 2.

Thoracic Tilt – Pelvis (projection as defined in Chapter 2): Extended PCA

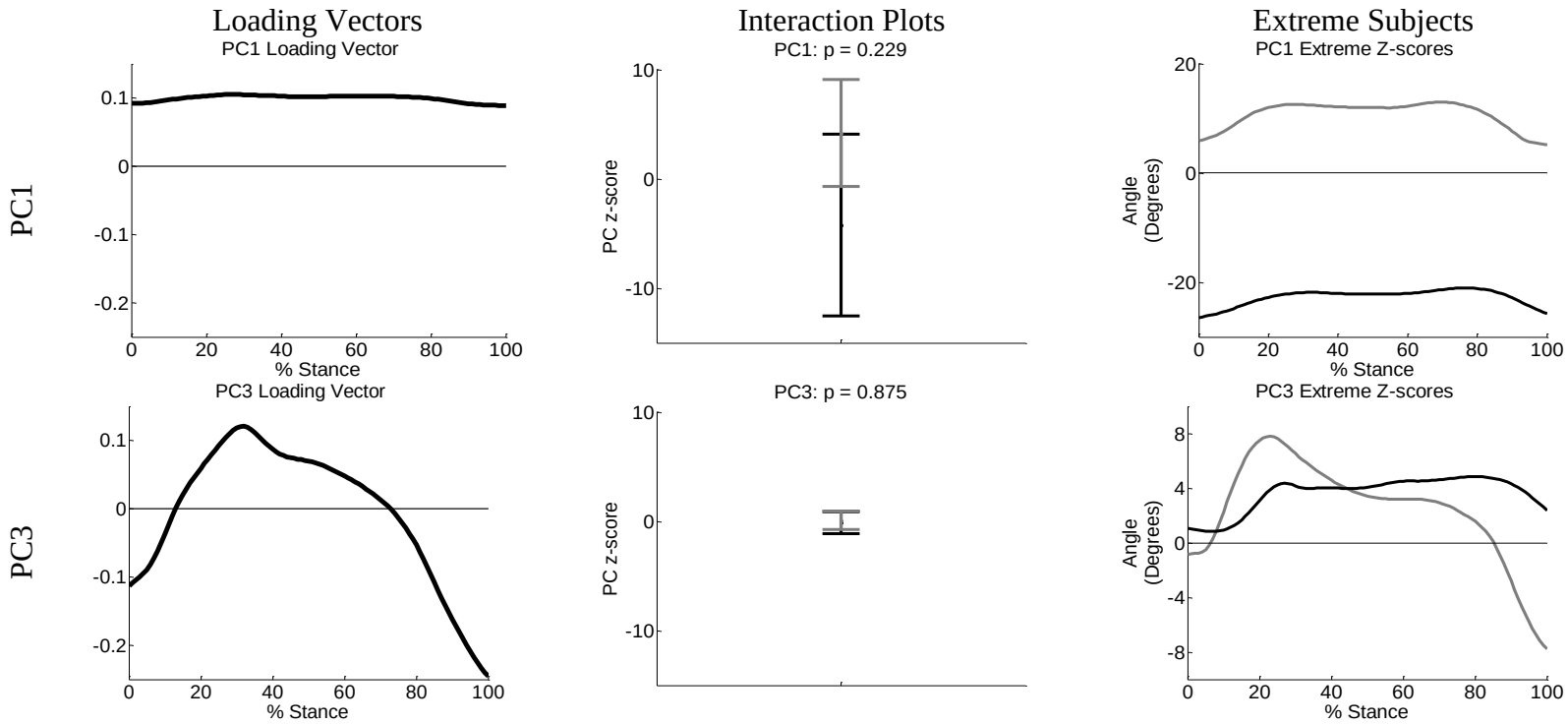

Figure C.2. PCA loading vectors, PCA interaction plots, and PCA extreme Z-scores for PC1 (top row) and PC3 (bottom row) for thoracic tilt – pelvis (projection) as defined in Chapter 2.

Thoracic Tilt – Lab (as defined in Appendix B): Extended PCA

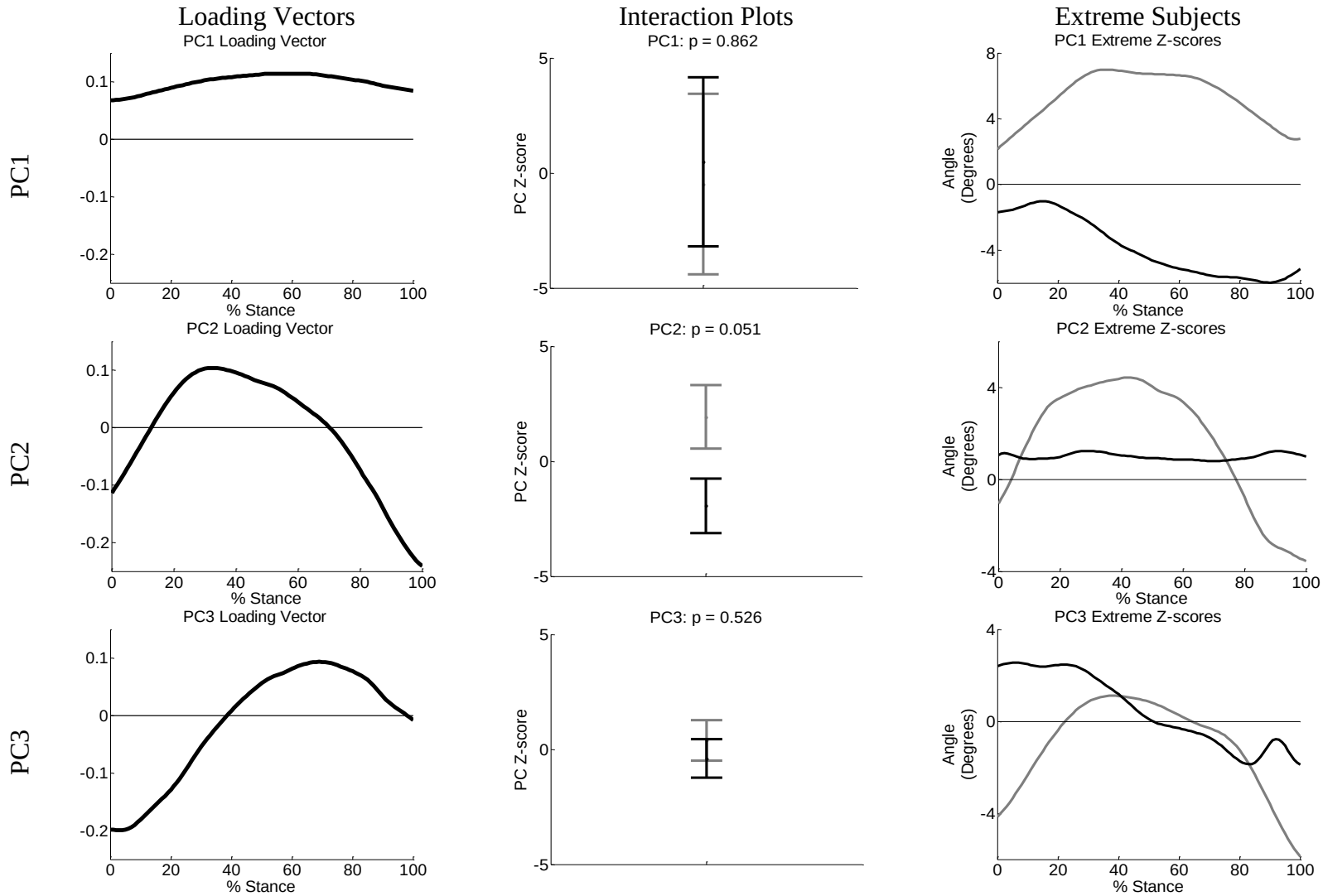

Figure C.3. PCA loading vectors, PCA interaction plots, and PCA extreme Z-scores for PC1 (top row) and PC3 (bottom row) for thoracic tilt – lab as defined in Appendix B.

Thoracic Tilt – Pelvis (as defined in Appendix B): Extended PCA

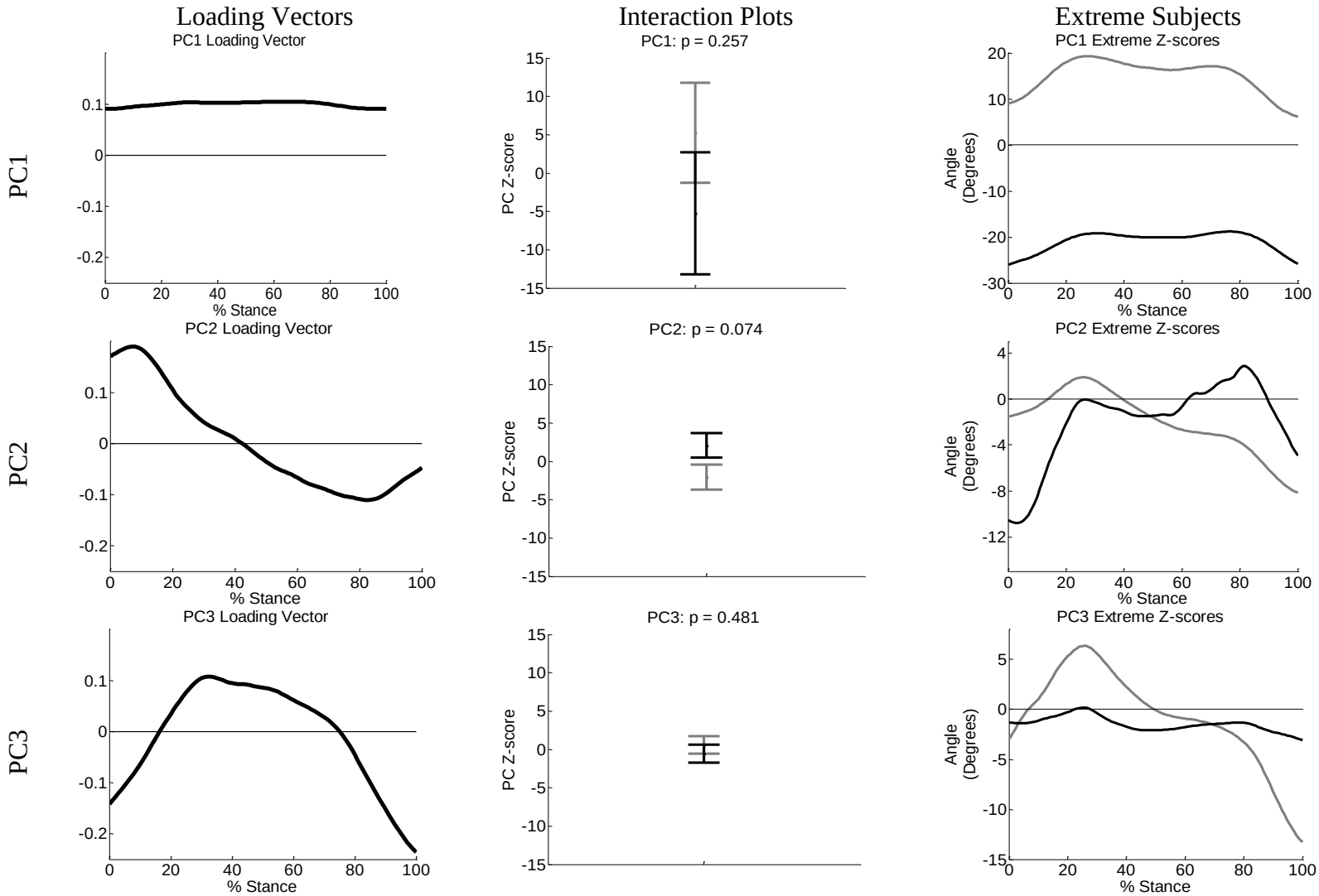

Figure C.4. PCA loading vectors, PCA interaction plots, and PCA extreme Z-scores for PC1 (top row) and PC3 (bottom row) for thoracic tilt – pelvis as defined in Appendix B.