THE EFFECTS OF ALTERNATIVE JOINT MODELS IN THE
STUDY OF LOWER-LIMB JOINT MOMENTS IN KNEE
OSTEOARTHRITIS

by

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Abstract

Knee osteoarthritis is a disease that affects nearly 40% of the global population over the age of seventy. It is believed that the incidence and progression of osteoarthritis can be partially attributed to changes in mechanical joint loading. Consequently, changes in lower-limb joint moments are important outcome measures for its treatment and prevention. The purpose of this study was to investigate the effect of four different anatomic joint models on the detection of changes in lower-limb joint moments due to knee osteoarthritis.

Moments during gait were calculated for 44 subjects with moderate osteoarthritis and 44 asymptomatic subjects, then expressed using four joint models: Joint Coordinate System, Plane of Progression, Distal, and Proximal. Discrete peak measures and principal component scores were compared between groups.

Hip adduction magnitude, knee adduction magnitude, peak early-stance knee internal rotation, and peak ankle plantarflexion moments were different between groups regardless of joint model. Differences detected using principal component analysis were less sensitive to the choice of joint model. Results support adoption of the Joint Coordinate System as a standard for joint moment expression due to its clinical relevance and ability to detect differences due to moderate knee osteoarthritis.
Co-Authorship

This thesis represents the original work of the author, Scott C. E. Brandon, completed under the supervision of Dr. Kevin J. Deluzio. Gait data were collected under Dr. Deluzio’s supervision at Dalhousie University in Halifax Nova Scotia, Canada. Joint moment data were extracted using custom software developed by Dr. Deluzio. The author was responsible for modeling, analysis, and interpretation of the data, as well as the writing of this document.
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To the JAM group: you made every week seem one day shorter, and led my accounts of the grad school experience to sound rather decadent. Thanks Simon Jones, Melanie Thompson, Andreas Burger, Stacey Acker, Miranda Restorick, Qingguo Li, and Nicole Badke for staying awake when I showed yet another diagram of joint moments, and for offering much-needed advice. Heather S. Linley, I think you deserve special thanks for the extra time you had to spend with me, and the number of times I interrupted your work, intentionally or otherwise. You showed me how a mathelete approaches engineering, and you were the true instigator for activities like pumpkin carving, office bread, the IT crowd and black books, and of course, Kevin’s april fools adventure.

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Chapter 1

Introduction

Osteoarthritis is a debilitating disease that can be found in any joint in the body, but is most common at the knee [1]. It is characterized physiologically by the loss of cartilage and hypertrophy of bone, and symptomatically by joint stiffness and pain [1]. The World Health Organization estimates that 40% of people over 70 years of age suffer from knee osteoarthritis, and as a result fully 25% of this population cannot perform major activities of daily life [1]. Osteoarthritis significantly reduces the quality of life of afflicted individuals, and its treatment places a great burden on the health care system.

Osteoarthritis severity is most commonly diagnosed in vivo using the radiographic Kellgren and Lawrence grading system [2]. Both the loss of articular cartilage volume, evidenced by joint space narrowing, and osteophyte formation can be detected radiographically as shown in Figure 1.1. To assess symptomatic disease severity, subjects can be graded using subjective surveys such as the Western Ontario and McMaster Universities (WOMAC) scale for the hip and knee, which is based on joint pain, stiffness, and function [3,4].
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Figure 1.1: Four stages of radiographic osteoarthritis
Joints are graded based on the presence five features: osteophyte formation, periarticular ossicles, joint space narrowing, sclerosis of subchondral bone, and altered bone end shape [2]. Gait is severely impaired in subjects with Grade 4 radiographic osteoarthritis.

Many risk factors have been implicated in the incidence and progression of knee osteoarthritis including genetics, obesity, aging, joint trauma, limb alignment, bone mineral density, physical activity, joint stability, muscle strength, and estrogen deficiency [5]. However, epidemiological studies of osteoarthritis have been limited by the lack of an objective diagnostic criterion that establishes a strong correlation between physical changes in the joint and clinical symptoms [5]. Subjects may exhibit joint degeneration or mechanical changes, yet report no change in physical function or pain, and vice versa. This dichotomy is present in the definition of osteoarthritis established at a workshop on osteoarthritic disorders in 1994 [6]:

Osteoarthritic diseases (OA) are a result of both mechanical and biological events that destabilize the normal coupling of degradation and synthesis of articular cartilage chondrocytes and extracellular matrix, and subchondral
bone. Although they may be initiated by multiple factors, including genetic, developmental, metabolic, and traumatic, OA diseases involve all of the tissues of the diarthrodial joint. Ultimately, OA diseases are manifested by morphologic, biochemical, molecular and biomechanical changes of both cells and matrix which lead to a softening, fibrillation, ulceration, loss of articular cartilage, sclerosis and eburnation of subchondral bone, osteophytes, and subchondral cysts. When clinically evident, OA diseases are characterized by joint pain, tenderness, limitation of movement, crepitus, occasional effusion, and variable degrees of inflammation without systemic effects.

In this definition, it is important to note that osteoarthritis is not referred to as a single entity, but rather as a conglomerate of various pathogenic characteristics from which it is difficult to establish causality [1]. There is nevertheless strong evidence supporting the role of mechanical factors in disease onset and progression.

1.1 Mechanical Factors and Osteoarthritis

It has been shown, using animal models, that osteoarthritis can be induced by altered mechanical loading of the knee joint [7,8]. Accordingly, studies have investigated the loading of the human tibiofemoral knee joint during activities of daily living in subjects with and without osteoarthritis. Chief among these activities in terms of repetitive loading of significant magnitude is walking, which is commonly segmented into the stance and swing phases as shown in Figure 1.2. During walking, or gait, subjects with osteoarthritis exhibit obvious changes such as reduced walking speed and swing-phase duration [9]. However, there are many other changes in both kinematics and kinetics,
summarized in Table 1.1, that are associated with the presence and progression of knee osteoarthritis.

Figure 1.2: The human gait cycle
Stance is when the foot of interest is on the ground, and swing is when the foot of interest is traveling through the air. For normal gait, the stance : swing ratio is approximately 60:40. Stance duration is increased in subjects with osteoarthritis [10].
Table 1.1: Summary of gait differences associated with knee osteoarthritis

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<td>Knee flexion angle</td>
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<tr>
<td>Range</td>
<td>OA &lt; Control</td>
<td>Andriacchi et al. (1982), Brinkmann et al. (1985), Messier et al. (1992), Stauffer et al. (1977)</td>
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<td>Peak during stance phase</td>
<td>OA &lt; Control</td>
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<td>OA &lt; Control</td>
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<td>Hip flexion angle</td>
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<td>Knee adduction moment</td>
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<tr>
<td>Peak</td>
<td>OA &gt;= Control †</td>
<td>Balunas et al. (2000), Gok et al. (2002), Weidenhohlm et al. (1994), Mundermann et al. (2005), Sharma et al. (1998), Kaufman et al. (2001), Landry et al. (2006)</td>
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<td>First Peak</td>
<td>OA &lt;= &gt; Control †</td>
<td>Deluzio et al. (2006), Hurwitz et al. (2002), Mundermann et al. (2005)</td>
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<td>Second Peak</td>
<td>OA &lt;= &gt; Control †</td>
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<td>Mid-stance</td>
<td>OA &gt; Control</td>
<td>Weidenhohlm et al. (1994)</td>
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<td>Overall stance-phase magnitude</td>
<td>OA &gt; Control</td>
<td>Deluzio et al. (2006), Diamond et al. (2005)</td>
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<td><strong>Knee flexion moment</strong></td>
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<td>Peak</td>
<td>OA &lt; Control</td>
<td>Balunas et al. (2000), Gok et al. (2002), McKeen et al. (2006)</td>
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<td>Overall stance-phase magnitude</td>
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<td>Deluzio et al. (2006)</td>
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<td>Early-stance magnitude</td>
<td>OA &lt; Control</td>
<td>Deluzio et al. (2006), Landry et al. (2006)</td>
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<td><strong>Knee extension moment</strong></td>
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<td>Early-stance magnitude</td>
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<td>Landry et al. (2006)</td>
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<td><strong>Hip flexion moment</strong></td>
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<td>Peak in late-stance</td>
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<td>Mundermann et al. (2005)</td>
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<td>Ankles inversion moment</td>
<td>OA &lt; Control</td>
<td>Mundermann et al. (2005)</td>
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<td><strong>Kinetics - Forces</strong></td>
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<td>Vertical ground reaction force</td>
<td>OA &lt;= &gt; Control †</td>
<td>Gok et al. (2002), Gyory et al. (1976), Stauffer et al. (1977), Cooke et al. (1997), Messier et al. (1992), Radin et al. (1991)</td>
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<tr>
<td>Bone-on-bone forces in frontal plane</td>
<td>OA &gt; Control</td>
<td>Teixeira et al. (1996)</td>
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Adapted from Tables 3-5 in "Techniques in Modern Gait Analysis and Their Application to the Study of Knee Osteoarthritis" [12].

† Multiple operators denote conflicting conclusions in literature.
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There is a conspicuous absence of joint contact force parameters in Table 1.1. This can be partially attributed to the challenge of analyzing an indeterminate system of stabilizing soft tissue surrounding the joint capsule, shown in Figure 1.3. It is, with current technology, impossible to accurately measure the forces in these tissues using non-invasive means.

*Figure 1.3: Knee anatomy*
There are many stabilizing ligaments and muscles acting on the knee joint. In advanced cases of osteoarthritis, articular cartilage becomes severely degraded and can even lead to exposed bone-on-bone contact [12].
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However, the changes in joint moments that have been found in the presence of knee osteoarthritis are related to both joint loading and muscle function [13]. An in vivo study of a patient with an instrumented prosthetic knee joint showed that the medio-lateral distribution of the contact force in the knee is directly correlated with the knee adduction moment [14]. Furthermore, differences in other lower-limb joint moments are important because they may reflect muscle strength deficiencies or other compensatory gait strategies [15]. Joint moments, obtained through gait analysis, can provide targets for clinical interventions [16-18] that will address joint contact forces and potentially reduce the incidence and rate of progression of knee osteoarthritis.

1.2 The Problem of Joint Models

Gait analysis is the standard for biomechanical evaluation of gait pathology. It involves the use of a motion capture system that is temporally and spatially aligned with an instrumented platform capable of recording six degree of freedom forces and moments [19]. Typically, each limb segment is assumed to be a rigid body with subject-specific mass and inertia properties [20]. The net force and moment applied externally to each joint can be calculated using an inverse dynamics algorithm based on Newton’s equations [21]. However, while gait analysis is a common procedure, many assumptions and techniques are not standardized across the research community, which can lead to conflicting results. In particular, there is no standard for the anatomical expression of net lower-limb joint moments.
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Schache and Baker [13] attempted to provide a definitive evaluation of the reference frames used to report lower-limb joint moments. They analyzed the gait of ten able-bodied subjects, and a single subject with cerebral palsy, and resolved moments into four alternative reference frames. Moments were amplitude-normalized to subject mass and time-normalized to percent of gait cycle. Comparing the magnitude and timing of local peak events in the moment waveforms, it was found that the choice of reference frame significantly affected many of the peak parameters. Moments about the long axis of each limb segment (internal rotation moments) were particularly sensitive to the choice of reference frame. Furthermore, the joint moments for the single subject with cerebral palsy suggested that abnormal gait kinematics would amplify the differences between reference frames. Thus, for studies of gait pathologies such as cerebral palsy or osteoarthritis the choice of reference frame is critical. While the authors recommended the use of the non-orthogonal Joint Coordinate System to agree with the standard for joint kinematics [22], in a later paper Schache et al. recognized the potential need for study-specific reference frames depending on the parameters and subjects of interest [23].

Newell et al. [24] also identified the need for standardization of the reference frame, or joint model, used to report the knee adduction moment in studies of knee osteoarthritis. A survey of recent literature revealed three conflicting models were used to extract a wide variety of parameters from the knee adduction moment waveforms. While group differences in knee adduction moments had been reported for subjects with osteoarthritis using all three joint models, the magnitude and location of these changes were not consistent across joint models. Analyzing the moments of 44 asymptomatic control
subjects and 44 subjects with knee osteoarthritis, it was found that the choice of joint model affected the moment waveforms differently for the osteoarthritis and control groups. The only discrete parameter that identified a difference between the two groups regardless of the choice of joint model was the local minimum magnitude at mid-stance. However, the authors also used principal component analysis, a multivariate statistical technique, to identify the magnitude of the moment across the entire stance phase as a second robust indicator of group separation. The peak adduction moment was not a reliable indicator of knee osteoarthritis across all joint models; this was an extremely important finding, considering its frequent use in studies of knee osteoarthritis. Furthermore, the test-retest reliability of the peak knee adduction moment has been investigated, but only within a single reference frame [25]. As the investigation of knee osteoarthritis expands from the adduction moment to other lower-limb joint moments [15], it is important to elucidate for all joints the confounding role of joint models in the detection of group differences.

1.3 Techniques for Data Reduction

Gait analysis produces a large volume of temporal data that must be reduced for interpretation. An example of the knee adduction moment data for eighty-eight subjects is plotted in Figure 1.4. While data reduction can be accomplished by extracting peak values from regions of interest, this method is limited in that it inherently assumes that each subject actually exhibits a given feature. Furthermore, if the region of interest is broad, disparate peaks may be drawn into comparison between subjects, complicating
CHAPTER 1: INTRODUCTION

interpretation of differences. This effect is exacerbated between studies for joint moments by the usage of alternative coordinate systems, which shift the location and ratio of peaks within the gait cycle [13,24]. Principal component analysis has been shown to objectively identify features of the gait waveforms that represent common modes of variation in amplitude, magnitude, and temporal synchronization between subjects [26]. In comparisons where the location of features within the gait cycle is important, PCA may provide a superior means of data reduction.

Figure 1.4: Raw knee adduction moment data
Data for 44 osteoarthritis and 44 control subjects are plotted versus percent of the stance phase. Note the appearance of group trends despite the large inter-subject variability.

1.4 Purpose

It is clear that the shape and magnitude of lower-limb joint moment waveforms are sensitive to the choice of joint model [13,24,27]. This makes it difficult to compare results between studies. However, even more important from the perspective of knee
CHAPTER 1: INTRODUCTION

osteoarthritis is the fact that subjects with abnormal gait kinematics express different changes in joint moments than able-bodied subjects depending on the choice of joint model [13,24]. Some reported differences between groups may actually be artifacts introduced by the use of a specific joint model. These differences might not be detected by research groups using alternative joint models. Conversely, some physiologically meaningful differences in joint moments may be masked by the choice of joint model. This uncertainty hinders the development, and inter-centre applicability, of clinical treatments based on joint moment parameters.

The issue of joint model selection has not been resolved for studies involving subjects with knee osteoarthritis. A greater understanding of the sensitivity of all lower-limb joint moments to the choice of joint model will be essential in adopting a standard model, and will provide valuable insight into the effects of knee osteoarthritis that are not model-dependent.
1.5 Objectives

1. To determine the effects of using four alternative joint models to analyze lower-limb joint moments in subjects with and without osteoarthritis.

2. To identify features of joint moment waveforms that change due to osteoarthritis regardless of the choice of joint model. Differences will be calculated using:
   - Discrete Peak Parameters: local max and min values obtained within a given window
   - Principal Component Analysis (PCA): a statistical method that restructures the data into features representing orthogonal components of variation

3. To identify the joint model that is most able to detect clinically meaningful changes in joint moments due to osteoarthritis.
1.6 Hypotheses

1. The choice of joint model affects the detection of differences in lower-limb joint moments between subjects with and without osteoarthritis.

2. Some joint moment waveform features will exhibit group differences regardless of the choice of joint model. These changes are robust measures of the effect of knee osteoarthritis that could be valuable outcome measures for clinical interventions.

3. Moments calculated using the Joint Coordinate System (JCS) model have the greatest physiological relevance. Thus, the JCS will be the model that is most able to detect changes in joint moments due to osteoarthritis.
Chapter 2

Background

2.1 Gait analysis

The primary element of a gait analysis laboratory is the motion capture system. Data in this study was collected using two Optotrak 3020 infrared cameras boasting an accuracy of 0.1mm in-plane and 0.15mm perpendicular to the viewing plane at a range of 2.25m [28,29]. Examining the measurement volume shown in Figure 2.1 it is clear that the subjects must walk much further from the camera to remain entirely in the viewing field. However, States et al. [29] showed that at the maximum distance from the camera the accuracy was only marginally diminished, and still much greater than alternative measurement systems. In this study, the measurement volumes of two Optotrak cameras were overlapped to increase the effective viewing field and to overcome line-of-sight restrictions of the optical tracking system.
Each Optotrak camera has three lenses to detect infrared light emitted from the markers, which are attached to the subject as shown in Figure 2.2. While only two lenses are required for triangulation, the third lens adds precision to the marker position estimates, particularly in the depth direction. For the purpose of gait analysis, each body segment is assumed to be rigid between joints. Segment motion can therefore be tracked relative to the global laboratory coordinate system in three dimensions using three or more non-collinear markers to create a segment-fixed coordinate system.
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2.1.1 Kinematics

The goal of gait analysis is to track bone and joint kinematics that represent a subject’s customary manner of walking. However, direct skeletal attachment of markers is a highly invasive procedure, and it may alter a subject’s gait. Alternatively, markers can be attached to the skin, but this introduces error caused by the motion of intermediate soft tissue between the markers and the underlying bones. Two proposed solutions include the use of redundant markers to allow a least-squares fit to the segment position, and the attachment of markers using rigid clusters [30]. The limitations of the redundant marker method are that it increases the computational cost of data processing, increases the time to set up the experiment, and can restrict the maximum sampling frequency. However, computational techniques that can accommodate deformable segment assumptions have shown significant improvements in accuracy through the use of redundant marker arrangements [31]. Rigid clusters are easier to mount and their geometric relationship can be exploited to reduce photogrammatic error [30]. In this study, rigid tracking clusters were located on thick neoprene bands as shown in Figure 2.2 to avoid areas of large skin motion [32] and to isolate the markers from some soft-tissue movement [30]. Clusters should be placed towards the lower (distal) thigh and upper (proximal) shank [32].
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Figure 2.2: Marker configuration on gait analysis subject
Tracking clusters of three markers (yellow dots) are shown on the foot, shank, thigh, and pelvis. Additional joint markers are used at the ankle, knee, hip, and shoulder.

To relate the coordinate system of the tracking clusters to an anatomical system embedded in the segment, the positions of landmarks on bone surfaces were taken relative to the tracking clusters during static calibration trials. These anatomical landmarks allow creation of basis vectors for anatomical coordinate systems that can be tracked, according to the rigid body assumption, relative to the cluster coordinate system using an invariant transformation matrix [33]. With anatomical axes thus prescribed, joint rotations can be calculated using a Cardan sequence recommended by the International Society of Biomechanics: flexion – adduction – internal rotation [22,34].
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Anatomical conventions for reporting joint rotations and moments are shown for the right leg in Figure 2.3. For the left leg, all flexion axes point in the same direction as shown, while internal rotation and adduction axes are reversed to maintain physiological meaning. The complete development of this non-orthogonal Joint Coordinate System from anatomical landmarks is described for each joint in Chapter 3.

Figure 2.3: Anatomical conventions for joint rotations and moments
Note: these are not orthogonal segmental coordinate systems. The names for mathematically negative rotations are given in brackets. Adduction and abduction are also known as varus and valgus directions, respectively.

It is important to note in these joint angle calculations that while the order of rotations is specified, no joint constraints are applied to the model. That is, joint angles are based only upon the relative orientation of the two adjacent segments at a given instant in time. There is no constraint on the translations and rotations allowed for each joint, or even for the length of each rigid segment. This simplification makes the model
CHAPTER 2: BACKGROUND

more susceptible to soft tissue motion artifact as unrealistic joint motions (leading to changing segment lengths and disconnected joints) are neither identified nor corrected [35]. However, it is difficult to characterize anatomical joints such as the knee and ankle which do not have a fixed centre or axis of rotation [36,37]. The simplified “inter-segment” joint angle method used in this study is in agreement with accepted standards for gait analysis [30,38], but is also subject to the same errors and limitations.

2.1.2 Kinetics

A summary of the inverse dynamics procedure for biomechanical analysis is given in Figure 2.4. Ground reaction forces and moments were measured using an AMTI force platform mounted in the same plane as the floor as shown in Figure 2.2. The origin of the laboratory coordinate system was registered to the centre of the force platform and the collection was synchronized with the Optotrak motion capture system; therefore, by calculating both the centre of pressure of the forces on the platform and the position of the body segments in the laboratory the recorded ground reaction forces can be applied to the subject model [38]. As gait involves relatively low segmental accelerations, a quasi-static method could be used to calculate joint forces and moments simply by taking the sum of forces and moments at the joint centres. However, increased accuracy can be obtained by incorporating the changes in momentum in the Newton-Euler equations [38]. This requires an estimation of segmental inertial parameters based on anthropometric regression equations [20]. Because this is a bottom-up inverse dynamics approach, the proximal (closest to the centre of the body) joint forces and moments will be most affected by the use of, and error in, inertial parameters [38]. Fortunately, as changes in
momentum are relatively small during gait, error in joint moments due to uncertainty in inertial parameters is much smaller than the error due to uncertainty in segment angles [39]. Lastly, it is important to recognize that because these are external moments and forces, they do not account for co-contraction of opposing muscle groups at a joint, therefore the joint forces in this model underestimate the true contact forces.

Figure 2.4: The inverse dynamics process
Note: Anatomical expression of joint moments and forces depends on the definition of the joint model, or coordinate system. This is not standardized in the literature.
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The external joint forces and moments must be expressed in local anatomic coordinate systems for physiological interpretation, based on the chosen joint model. Early gait studies employed a planar approach that assumed a global flexion axis for all subjects; however, incorporating a three dimensional model provides valuable information [40]. There are essentially five alternative coordinate systems that could be used to express joint moments. The first is the global laboratory system, based in the force platform [41]. However, beyond the fact that subjects walk in a plane approximately perpendicular to one of the primary axes in this frame, it has little physiological relevance and was therefore not considered in this study. The second coordinate system is a planar representation that allows the vertical (internal rotation) axis and the longitudinal (adduction) axis to follow the rotation of the distal segment while maintaining the global flexion axis [42]. The remaining three alternatives are three dimensional systems based in either the proximal (above the joint) segment, the distal (below the joint) segment, or a combination of the both. Taking one axis from each of the proximal and distal segments and calculating an intermediate floating axis results in the Joint Coordinate System introduced in section 2.2.1 and described in Chapter 3.
2.2 Principal Component Analysis

Principal component analysis (PCA) is a statistical technique that transforms a multivariate dataset into an uncorrelated set of components [43]. The process could be compared to performing a fourier transformation on a signal to identify its constituent components as a linear combination of sine and cosine waves of various frequencies and phase shifts. In biomechanical signals, there are certainly sinusoidal features that could be extracted using fourier analysis. However, with PCA the constituent components are not assumed a priori, but rather they are the eigenvectors of the data. This reduces the number of unique components required to characterize the majority of variation in the data [44]. A limitation of this method is that independent studies will always develop slightly different components, therefore numeric results are difficult to compare in absolute terms. Nevertheless, PCA has shown a unique ability to detect correlated changes in magnitude, amplitude, and timing throughout the gait cycle [45]. Finally, as the knowledge of normal and pathological gait patterns grows, it would be conceivable to develop characteristic principal component curves for standardized comparisons between laboratories.

2.2.1 PCA Procedure

Briefly, to perform PCA the gait data was normalized to 101 points representing each percentage of the stance phase of the gait cycle for each subject. The data was mean-centred, then the covariance matrix was calculated. Next, the eigenvectors of the
covariance matrix were extracted and ranked according to decreasing eigenvalue. Finally, z-scores were calculated for each subject for each eigenvector [43]. Z-scores are non-dimensional weight factors representing the degree to which a subject expresses the characteristic described by the eigenvector, or loading vector.

2.2.2 Data Reduction

As each subject gets an individual z-score for each loading vector, the maximum number of components would be equal to the lesser of the original number of observations or the number of subjects. In this case, the original dimensionality of the dataset would be retained despite being restructured into orthogonal components. Fortunately, about 90% of the variation in gait waveform data can be explained by the first three principal components [44] and the remainder of the principal components describe individual subject variability and smaller group effects. It is important to note that principal components explaining very small percentages of the total variability can be among the most discriminatory factors [43]. However, it can be difficult to attach clinical significance to such small features. Thus, PCA reduces the data from 101 correlated values for each subject to about three independent parameters without relying upon subjective identification of relevant features.

2.2.3 Physiological Interpretation

Any objective reduction in the dimensionality of gait data is irrelevant if the resultant parameters cannot be related to physical changes. Indeed, the challenge of interpretation is a barrier to both communication of results and wide-spread usage of PCA in the gait
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community. In this study, a combination of graphical methods was used to interpret each principal component: loading vectors, extreme subjects, and reconstructed curves. While graphical methods could be criticized for requiring subjective interpretation of “objective” results, it is the most intuitive for communication. Furthermore, if the changes cannot be depicted graphically, they probably have no clinical relevance.

2.2.4 Loading Vectors

The eigenvectors are often called loading vectors [44,45], or characteristic vectors [43]. They represent correlated deviations from the mean throughout the gait cycle. For example, if subjects that tend to have higher peaks at 25% stance also tend to have lower peaks at 75% stance, and both of these changes are independent of other modes of variation, then one loading vector will appear as a sinusoid, or amplitude feature, that captures both of these peak changes. Examples of typical loading vectors for amplitude, magnitude, and phase-shift effects are shown in Figure 2.5 a). A zero-line is helpful in identifying whether a positive z-score applied to the loading vector will have an additive or subtractive effect on the mean curve. The amplitude operator is found when opposite peaks in the loading vector coincide with the locations of the peaks in the original data. For the example shown, a positive z-score would denote a higher first peak and lower second peak. The magnitude operator is the easiest to interpret as it only has a single x-intercept throughout the majority of the gait cycle. However, the phase shift operator appears to be very similar to an amplitude operator. The only difference is that in this case the peaks in the original data (not shown) coincided with the x-intercepts between peaks in the loading vector. Thus, this principal component adds to one side of the peak
and subtracts from the other, effectively introducing a phase shift. It is clear that the loading vectors must be interpreted in light of the original data and the other graphical methods.

Figure 2.5: Principal component interpretation
Columns are examples of the three graphical tools for interpreting amplitude, magnitude, and phase shift effects commonly found in joint moment data. a) Loading vectors: the eigenvectors of the gait cycle waveform, showing modes of variation within the original data. b) Extreme subject waveforms: original data for subjects with high (solid black) and low (dashed grey) z-scores. c) Single-component reconstructed curves: mean waveforms for groups with high and low z-scores showing the between-groups effect of the principal component.
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2.2.5 Extreme Subjects

Another interpretive tool is a plot of the original data for subjects who received high and low (extreme) z-scores for a given loading vector. Differences between the high and low subjects can be attributed to the features captured by the loading vector, as shown in Figure 2.5 b). Plots of extreme subjects are highly valuable because they characterize the effect of a loading vector in the original units and demonstrate the range of observations in the population. In the examples selected for Figure 2.5 b), the separation between high and low z-scores is relatively clear. However, it is important to plot more than a single subject from each extreme to ensure that the graphical separation is not simply the result of outlier subjects. There is often a large amount of variability in the extreme subject plots because a subject can have extreme characteristics that were captured by more than one loading vector. Thus, to ensure the interpretation is capturing only the effects of the selected principal component, it is therefore helpful to consider a third graphical representation: single component reconstructions.

2.2.6 Reconstructed Curves

An important property of principal component analysis is that the eigenvector matrix is orthonormal, therefore the principal component model can be inverted and the original data can be expressed in terms of a linear combination of the principal components [43]. This method can be used to demonstrate the convergence of the true group means with means constructed using only the first few principal components. A general equation for reconstructing a single subject’s waveform from z-scores and loading vectors is given in
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Equation 1.

\[ x_i = \bar{x} + u_1 * z_{1i} + u_2 * z_{2i} + u_3 * z_{3i} + \cdots + u_k * z_{ki} \]

Where: 
- \( x_i \) = observation for single subject \( i \)
- \( u_j \) = loading vector for principal component \( j \)
- \( z_{ij} \) = z-score for subject \( i \), principal component \( j \)
- \( \bar{x} \) = mean of all subjects
- \( k \) = number of PC’s retained

From this equation, it can be seen that reconstruction of the original data involves adding the contribution of each principal component to the mean of the original data. Thus, it is possible to isolate the contribution of a single principal component to a subject’s waveform by adding only that principal component to the original mean. Figure 2.5 c) shows the separation between the means of two groups, reconstructed using only a single principal component and the mean of the original data. Taking the means of reconstructed waveforms from two groups provides a visualization of the between-groups effect size for a given principal component. This difference is expressed in terms of actual physiological units. Compared to the plot of extreme subject waveforms in Figure 2.5 b), the reconstructed waveforms are much smoother because they contain only the variation due to a specific loading vector.
Chapter 3

The Effect of Joint Models on Differences in Joint Moments due to Knee Osteoarthritis

A paper submitted to *Gait & Posture* in May, 2009

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Kevin J. Deluzio
CHAPTER 3: THE EFFECT OF JOINT MODELS

3.1 Introduction

Subjects with medial compartment knee osteoarthritis have been shown to exhibit changes in joint moments at the ankle, knee, and hip [15,46-48]. These changes are considered to be risk factors for disease progression [49,50] and have been implicated as predictors of surgical outcome [18]. Accordingly, lower-limb joint moments have been used to measure the effectiveness of gait interventions including toe-out walking [51], trunk lean [52], medial thrust gait [16], knee braces [17], and lateral heel wedges [53].

As the study of knee osteoarthritis moves beyond characterization of pathomechanics to the development of clinical interventions, it is important to ensure that outcome measures are robust. However, studies have shown that lower-limb joint moments exhibit systematic variation due to the differing joint models, or reference frames, used to resolve moments along anatomically meaningful axes. Schache and Baker [13] found that discrete measures of asymptomatic hip, knee, and ankle moments changed significantly when expressed in four alternative reference frames. They also demonstrated, using a single subject with cerebral palsy, that the choice of joint model will have a different effect on joint moments in groups with atypical kinematics as compared to asymptomatic controls. For subjects with medial knee osteoarthritis, Newell et al. [54] showed that the choice of joint model influences the detection of differences in discrete peak measures of knee adduction moments.

While the Joint Coordinate System developed by Grood and Suntay [55] is widely accepted as the standard for expression of lower-limb joint kinematics, there is no
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standard model for the expression of joint moments [23]. Two anatomic joint models that are commonly used in studies of knee osteoarthritis are the Distal model [46,56] and the non-orthogonal Joint Coordinate System [48,57]. A third anatomic joint model, referred to in this paper as the Plane of Progression (PoP) model, combines the global mediolateral axis with a projection of the distal internal rotation and adduction axes onto the Plane of Progression [15,58]. Moments can also be expressed in the Proximal anatomic model, which shares its flexion axis with the Joint Coordinate System [13]. The global laboratory reference frame has been considered as an alternative for joint moment expression [13,41], but to our knowledge it has not been used in studies of knee osteoarthritis.

The most common outcome measure extracted from a joint moment waveform is a discrete measure of peak magnitude. However, the model-dependency of peak measures identified by Newell et al. [54], as well as the subjective nature of peak detection, is a source of concern. Principal component analysis has been shown to objectively identify overall waveform features that differentiate between subjects with and without knee osteoarthritis [26]. In addition, Newell et al. [54] found that some features of knee adduction moment waveforms extracted using principal component analysis were insensitive to changes in joint models, unlike discrete peak parameters. While discrete peak measures are easy to extract, it may be necessary to use an alternative method of multivariate analysis such as principal component analysis in order to obtain robust measures in studies of human gait.
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The purpose of this study was to investigate the effect of joint model selection on the detection of changes in lower-limb joint moments due to osteoarthritis. We hypothesized that: i) group differences in some joint moments would be affected by the choice of joint model and ii) there exist some robust waveform measures that show changes due to osteoarthritis regardless of the choice of joint model.

3.2 Methods

3.2.1 Subjects

Forty-four subjects with radiographic medial knee osteoarthritis (Kellgren/Lawrence ≤ 3) were recruited from the Orthopaedic and Sports Medicine Clinic of Nova Scotia. Forty-four asymptomatic subjects were recruited for the control group through postings on the Dalhousie University campus. All subjects gave informed consent and the study was approved by the Institutional Review Board. Subjects were excluded if they had any other forms of arthritis, neuromuscular disorders, trauma or major surgery to the lower limb, or a history of stroke or cardiovascular disease. Symptomatic subjects were assessed using the Western Ontario and McMaster Universities (WOMAC) and radiographic Kellgren-Lawrence (KL) grading scales.

3.2.2 Gait

Subjects performed at least 4 (median = 6, 5th-95th percentile range = 4-8) walking trials at self-selected speed in their own low-top walking shoes. Kinematic data were collected for the affected limb at 100Hz using an Optotrak 3D motion analysis system.
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(Northern Digital Inc., Waterloo, ON). Rigid clusters of three markers were placed on the pelvis, thigh, shank, and foot in addition to single markers placed over the greater trochanter, lateral epicondyle, and lateral malleolus. The right and left anterior superior iliac spine (ASIS), medial epicondyle, fibular head, tibial tuberosity, medial malleolus, second metatarsal head, and calcaneus were digitized as virtual points to allow the creation of anatomical reference frames [59]. Unilateral ground reaction forces and moments were collected at 1000Hz using an AMTI force platform (Advanced Mechanical Technology Inc., Watertown, MA).

3.2.3 Segment Coordinate Systems

Ankle and knee joint centers were defined as the midpoint of the inter-malleolar and inter-epicondylar axes, respectively. Hip joint centers were defined using regression equations [60]. Anatomic coordinate systems for each segment were constructed from the locations of digitized virtual points and tracking markers using the right hand rule.

The lateral-medial axis for the pelvis segment was defined from the right ASIS to the left ASIS. The inferior-superior axis was mutually perpendicular to this lateral-medial axis and a vector from the cluster marker at the sacrum to the right ASIS. The posterior-anterior axis was mutually perpendicular to the other two axes.

For the thigh segment, the lateral-medial axis was defined from the lateral epicondyle to the medial epicondyle. The posterior-anterior axis was mutually perpendicular to this lateral-medial axis and a vector from the lateral epicondyle to the greater trochanter. The inferior-superior axis was mutually perpendicular to the other two axes.
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The lateral-medial axis for the shank segment was defined from the lateral malleolus to the medial malleolus. The posterior-anterior axis was mutually perpendicular to this lateral-medial axis and a vector from the lateral malleolus to the fibular head. The posterior-anterior axis was mutually perpendicular to the other two axes.

For the foot segment, the anterior-posterior axis was defined from the calcaneus to the second metatarsal head. The inferior-superior axis was mutually perpendicular to the anterior-posterior axis and a vector from the lateral malleolus to the medial malleolus. The lateral-medial axis was mutually perpendicular to the other two axes.

3.2.4 Joint Models for expression of Joint Moments

Four alternative joint models are shown in Figure 3.1: Joint Coordinate System (JCS), Plane of Progression (PoP), Distal, and Proximal models. The PoP model fixes the flexion axis perpendicular to the Plane of Progression. The internal rotation axis becomes a projection of the inferior-superior axis of the distal segment onto the plane of progression. The adduction axis is the mutual perpendicular of the flexion and internal rotation axes. The Distal model uses all three axes of the anatomic frame in the distal segment as the flexion, adduction, and internal rotation axes. The Proximal model uses the axes from the proximal segment. The Joint Coordinate System uses the proximal medial-lateral axis as the flexion axis, the distal inferior-superior axis as the internal rotation axis, and then creates a mutually perpendicular floating axis for the adduction axis. Because the proximal flexion and distal internal rotation axes are not perpendicular, this is a non-orthogonal coordinate system.
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Figure 3.1: Four alternative joint models
Models are shown using the knee for example. The Plane of Progression (PoP) model fixes the flexion axis perpendicular to the plane of progression, and the other two axes follow the distal segment, projected onto this plane. The Distal model follows the distal segment in three dimensions, while the Proximal model follows the proximal segment. The Joint Coordinate System (JCS) model uses the flexion axis from the Proximal model, the internal rotation axis from the Distal model, and a floating adduction axis, FA, that is perpendicular to both.

3.2.5 Data Analysis

Marker coordinate data were smoothed using a Butterworth filter with 10Hz cutoff frequency. Joint kinematics were calculated using the Joint Coordinate System [55]. Net lower limb external joint moments were calculated in the laboratory reference frame using a standard inverse dynamics approach implemented in MATLAB (The MathWorks, Natick, MA, USA). Segment inertial parameters were estimated from regression equations based on anthropometric measures [61]. Joint moments were time normalized to 101 points representing each percent of the stance phase of gait using cubic spline interpolation. Moments were also normalized to body mass, then resolved into components along anatomic axes in each of the four alternative joint models: JCS, PoP,
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Distal, and Proximal. The mean moment was obtained for each subject across all walking trials.

Local maxima and minima were extracted for analysis for each moment. Due to inter-subject and inter-group waveform variability, peak values for each subject were taken within a window of ± 4% stance from the mean peak location across all subjects. The window size was selected to ensure that a local peak value was selected for each subject; that is, a larger window did not affect magnitude of the discrete parameter. In addition to discrete parameters, principal component analysis was used to extract waveform features representing changes in magnitude and amplitude that capture more completely the differences between subjects with and without osteoarthritis [44]. Principal component analysis yields a loading vector for each type of variation within the data, and a set of z-scores corresponding with the degree to which each subject expresses the feature described by the loading vector.

3.2.6 Statistics

A two-way repeated measures ANOVA was implemented in MATLAB (The MathWorks, Natick, MA, USA) to test for significant group, model, and group-model interaction effects in both discrete parameters and principal component z-scores using a significance level of $\alpha = 0.05$ [62]. Joint model was the repeated measure. Post-hoc Tukey tests were used to investigate pair-wise group comparisons for each level of joint model.
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3.3 Results

Subjects with osteoarthritis were taller, heavier, older, and had a greater BMI than control subjects, as shown in Table 3.1. There was no difference in walking speed between the two groups. Osteoarthritis subjects had mean ± SD WOMAC pain scores of 7 ± 4 and function scores of 23 ± 13. All Osteoarthritis subjects demonstrated moderate radiographic medial knee osteoarthritis with KL scores between 1 and 3.

<table>
<thead>
<tr>
<th>Table 3.1: Subject demographics and walking speeds: mean (SD)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Gender</td>
</tr>
<tr>
<td>--------</td>
</tr>
<tr>
<td>Control (N = 44)</td>
</tr>
<tr>
<td>OA (N = 44)</td>
</tr>
</tbody>
</table>

* indicates statistical difference (p < 0.05) between control and osteoarthritis (OA) groups.

Mean lower-limb external joint moments for both control and osteoarthritis groups are shown with shaded standard deviations in Figures 3.2-3.4. Hip internal rotation, knee internal rotation, ankle adduction, and ankle eversion moments were highly dependent on the choice of joint model. Other moments were less sensitive to joint model, but still exhibited changes in both magnitude and shape. Changing the joint model affected the control and osteoarthritis subjects differently, as evidenced by the disappearance of group separation in distal model hip flexion moments and proximal model hip internal rotation moments.
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Figure 3.2: Mean external hip joint moments

Results are plotted for control (solid blue line) and osteoarthritis (dashed red line) groups at self-selected walking pace, expressed using four alternative joint models. Shaded regions are standard deviation bands for control (light blue) and osteoarthritis (light red) groups. Purple shading represents overlapping standard deviations. Note: the Joint Coordinate System uses the same flexion axis as the Proximal model and the same internal rotation axis as the Distal model.
Figure 3.3: Mean external knee joint moments
Results are plotted for control (solid blue line) and osteoarthritis (dashed red line) groups at self-selected walking pace, expressed using four alternative joint models. Shaded regions are standard deviation bands for control (light blue) and osteoarthritis (light red) groups. Purple shading represents overlapping standard deviations. Note: the Joint Coordinate System uses the same flexion axis as the Proximal model and the same internal rotation axis as the Distal model.
Figure 3.4: Mean external ankle joint moments
Results are plotted for control (solid blue line) and osteoarthritis (dashed red line) groups at self-selected walking pace, expressed using four alternative joint models. Shaded regions are standard deviation bands for control (light blue) and osteoarthritis (light red) groups. Purple shading represents overlapping standard deviations. Note: the Joint Coordinate System uses the same flexion axis as the Proximal model and the same internal rotation axis as the Distal model.
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3.3.1 Discrete Peak Parameters

Each of the nineteen discrete peak parameters demonstrated a significant model effect ($p < 0.05$) representing changes in moment magnitude between models, regardless of the group. Group differences, shown in Table 3.2, were present in all three planes for the hip and knee, but only in the sagittal plane at the ankle. Only four discrete peak parameters were different between groups regardless of the choice of joint model: late-stance hip adduction, mid-stance knee adduction, early-stance knee internal rotation, and late-stance ankle plantarflexion moments.

There was no group difference in either the early-stance or late-stance knee adduction moment peak. Early-stance group differences in hip internal rotation moments and knee flexion moments were found only in the PoP model, while late-stance knee internal rotation moments were different only in the Proximal model. The Distal was the only model where early-stance hip adduction moments were not significantly different.
### CHAPTER 3: THE EFFECT OF JOINT MODELS

#### Table 3.2: Discrete peak parameters [Nm/kg]: mean (SD)

<table>
<thead>
<tr>
<th>Gait measure</th>
<th>Feature</th>
<th>Location</th>
<th>Group</th>
<th>JCS</th>
<th>Poi</th>
<th>Distal</th>
<th>Proximal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Adduction Moment</td>
<td>Max</td>
<td>Early</td>
<td>Control</td>
<td>1.42 (0.20)</td>
<td>1.45 (0.19)</td>
<td>1.43 (0.22)</td>
<td>1.38 (0.15)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>1.29 (0.31)</td>
<td>1.32 (0.32)</td>
<td>1.33 (0.34)</td>
<td>1.23 (0.30)</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>Mid</td>
<td>Control</td>
<td>0.89 (0.15)</td>
<td>0.89 (0.15)</td>
<td>0.87 (0.15)</td>
<td>0.86 (0.15)</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.86 (0.25)</td>
<td>0.97 (0.25)</td>
<td>0.96 (0.25)</td>
<td>0.93 (0.24)</td>
</tr>
<tr>
<td></td>
<td>Max</td>
<td>Late</td>
<td>Control</td>
<td>1.34 (0.17)</td>
<td>1.34 (0.17)</td>
<td>1.33 (0.17)</td>
<td>1.33 (0.17)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>1.14 (0.28)</td>
<td>1.14 (0.28)</td>
<td>1.12 (0.28)</td>
<td>1.13 (0.28)</td>
</tr>
<tr>
<td>Flexion Moment *</td>
<td>Max</td>
<td>Early</td>
<td>Control</td>
<td>0.55 (0.19)</td>
<td>0.55 (0.19)</td>
<td>0.58 (0.24)</td>
<td>0.57 (0.17)</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.62 (0.27)</td>
<td>0.59 (0.31)</td>
<td>0.72 (0.26)</td>
<td>0.69 (0.26)</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>Late</td>
<td>Control</td>
<td>-0.10 (0.22)</td>
<td>0.05 (0.29)</td>
<td>-0.15 (0.20)</td>
<td>0.00 (0.19)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.00 (0.19)</td>
<td>0.05 (0.27)</td>
<td>-0.02 (0.21)</td>
<td>0.17 (0.21)</td>
</tr>
<tr>
<td>Internal Rotation Moment *</td>
<td>Min</td>
<td>Early</td>
<td>Control</td>
<td>-0.34 (0.09)</td>
<td>-0.34 (0.10)</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>-0.27 (0.13)</td>
<td>-0.30 (0.14)</td>
<td>0.05</td>
<td>0.05</td>
</tr>
<tr>
<td></td>
<td>Max</td>
<td>Early</td>
<td>Control</td>
<td>-0.12 (0.10)</td>
<td>0.12 (0.11)</td>
<td>0.22 (0.19)</td>
<td>0.20 (0.16)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.10 (0.11)</td>
<td>0.09 (0.12)</td>
<td>0.02 (0.12)</td>
<td>0.16 (0.15)</td>
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<tr>
<td>Knee</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Adduction Moment</td>
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<td>Control</td>
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<td>0.49 (0.11)</td>
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<td>OA</td>
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<td>0.47 (0.16)</td>
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<tr>
<td></td>
<td>Min</td>
<td>Mid</td>
<td>Control</td>
<td>0.23 (0.08)</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.31 (0.15)</td>
<td>0.32 (0.15)</td>
<td>0.33 (0.14)</td>
<td>0.32 (0.15)</td>
</tr>
<tr>
<td></td>
<td>Max</td>
<td>Late</td>
<td>Control</td>
<td>0.33 (0.10)</td>
<td>0.37 (0.09)</td>
<td>0.43 (0.09)</td>
<td>0.34 (0.10)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.37 (0.15)</td>
<td>0.39 (0.13)</td>
<td>0.45 (0.14)</td>
<td>0.38 (0.15)</td>
</tr>
<tr>
<td>Flexion Moment *</td>
<td>Max</td>
<td>Early</td>
<td>Control</td>
<td>0.48 (0.15)</td>
<td>0.55 (0.15)</td>
<td>0.47 (0.18)</td>
<td>0.47 (0.18)</td>
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<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.37 (0.27)</td>
<td>0.47 (0.28)</td>
<td>0.39 (0.29)</td>
<td>0.39 (0.29)</td>
</tr>
<tr>
<td></td>
<td>Min</td>
<td>Late</td>
<td>Control</td>
<td>-0.36 (0.12)</td>
<td>-0.32 (0.12)</td>
<td>-0.40 (0.12)</td>
<td>-0.36 (0.12)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>-0.34 (0.22)</td>
<td>-0.18 (0.21)</td>
<td>-0.36 (0.22)</td>
<td>-0.36 (0.22)</td>
</tr>
<tr>
<td>Internal Rotation Moment *</td>
<td>Min</td>
<td>Early</td>
<td>Control</td>
<td>-0.06 (0.06)</td>
<td>-0.06 (0.06)</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>-0.04 (0.06)</td>
<td>-0.11 (0.08)</td>
<td>0.05</td>
<td>0.05</td>
</tr>
<tr>
<td></td>
<td>Max</td>
<td>Early</td>
<td>Control</td>
<td>0.04 (0.03)</td>
<td>0.11 (0.03)</td>
<td>0.11 (0.03)</td>
<td>0.04 (0.03)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.05 (0.04)</td>
<td>0.05 (0.04)</td>
<td>0.05 (0.04)</td>
<td>0.05 (0.04)</td>
</tr>
<tr>
<td></td>
<td>Max</td>
<td>Late</td>
<td>Control</td>
<td>0.12 (0.04)</td>
<td>0.20 (0.05)</td>
<td>0.07 (0.05)</td>
<td>0.12 (0.12)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.12 (0.04)</td>
<td>0.24 (0.05)</td>
<td>0.22 (0.05)</td>
<td>0.12 (0.12)</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Adduction Moment</td>
<td>Max</td>
<td>Early</td>
<td>Control</td>
<td>0.28 (0.14)</td>
<td>0.06 (0.04)</td>
<td>0.24 (0.20)</td>
<td>0.33 (0.13)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>0.27 (0.14)</td>
<td>0.06 (0.03)</td>
<td>0.22 (0.21)</td>
<td>0.31 (0.15)</td>
</tr>
<tr>
<td>Plantarflexion Moment *</td>
<td>Min</td>
<td>Late</td>
<td>Control</td>
<td>-1.43 (0.12)</td>
<td>-1.41 (0.11)</td>
<td>-1.37 (0.11)</td>
<td>-1.37 (0.11)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>-1.31 (0.30)</td>
<td>-1.28 (0.29)</td>
<td>-1.25 (0.28)</td>
<td>-1.25 (0.28)</td>
</tr>
<tr>
<td>Eversion Moment *</td>
<td>Min</td>
<td>Late</td>
<td>Control</td>
<td>-0.08 (0.07)</td>
<td>-0.08 (0.07)</td>
<td>-0.28 (0.12)</td>
<td>-0.28 (0.12)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>OA</td>
<td>-0.06 (0.07)</td>
<td>-0.25 (0.15)</td>
<td>0.48</td>
<td>0.48</td>
</tr>
</tbody>
</table>

*p*-values (italic) obtained from Tukey post-hoc test following 2-factor ANOVA with joint model as the repeated measure. **Bold** denotes statistical significance (p < 0.05). Location refers to the portion of stance where the feature is found. † † feature of interest was not found in this model.

* Note that the JCS model uses the same flexion axis as the Proximal model and the same internal rotation axis as the Distal model. The JCS model was therefore omitted from the analysis of inter-model differences for these moments.

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3.3.2 Principal Component Analysis

Waveform features that differentiated between control and osteoarthritis groups are shown in Table 3.3. There was a significant model effect ($p < 0.05$) for each of the nine principal component; moments were different between joint models, regardless of the group, for every measure of waveform magnitude and amplitude.

The interpretation of the first principal component for the hip flexion moment is shown in Figure 3.5. The positive loading vector represents a magnitude shift proportional to each subject’s z-score. The group difference is significant in the Proximal/JCS model, but not in the Distal or the PoP ($p = 0.07$) models. Extreme subjects, taken from the upper and lower deciles of the z-scores, show the effect of this feature in terms of actual moments.
Table 3.3: Principal Component Analysis

<table>
<thead>
<tr>
<th>Gait measure</th>
<th>PC</th>
<th>Group</th>
<th>Model</th>
<th>Interaction</th>
<th>Interpretation: OA subject trends</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Adduction Moment</td>
<td>1</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
<td>0.12</td>
<td>Decreased magnitude throughout stance</td>
</tr>
<tr>
<td></td>
<td>3</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
<td>0.06</td>
<td>Increased ratio of mid-stance to early/late stance magnitude</td>
</tr>
<tr>
<td>Flexion Moment *</td>
<td>1</td>
<td>0.17</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
<td>Increased magnitude throughout stance (only Proximal model)</td>
</tr>
<tr>
<td>Internal Rotation Moment *</td>
<td>2</td>
<td>0.16</td>
<td>&lt;0.01</td>
<td>0.94</td>
<td>Decreased amplitude</td>
</tr>
<tr>
<td>Kne</td>
<td>Adduction Moment</td>
<td>1</td>
<td>0.03</td>
<td>&lt;0.01</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>0.08</td>
<td>&lt;0.01</td>
<td>0.97</td>
<td>Decreased amplitude</td>
</tr>
<tr>
<td></td>
<td>2</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
<td>0.3</td>
<td>Decreased amplitude</td>
</tr>
<tr>
<td>Ankle</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Adduction Moment</td>
<td>3</td>
<td>&lt;0.01</td>
<td>&lt;0.01</td>
<td>0.36</td>
<td>Increased magnitude 0-50% stance</td>
</tr>
<tr>
<td>Plantarflexion Moment *</td>
<td>1</td>
<td>0.07</td>
<td>&lt;0.01</td>
<td>0.91</td>
<td>Increased amplitude</td>
</tr>
<tr>
<td>Eversion Moment *</td>
<td>2</td>
<td>0.09</td>
<td>&lt;0.01</td>
<td>0.08</td>
<td>Increased magnitude 0-70% stance (not in Proximal model)</td>
</tr>
</tbody>
</table>

*p-values (italic)* for group, model, and interaction effects from a 2-factor ANOVA with joint model as the repeated measure.

**Bold** denotes statistical significance (*p < 0.05*).

* Note that the JCS model uses the same flexion axis as the Proximal model and the same internal rotation axis as the Distal model.

The JCS model was therefore omitted from the analysis of inter-model differences for these moments.
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Figure 3.5: Model-dependent hip flexion moment magnitude difference
Control (solid black line) and osteoarthritis (dashed grey line) extreme subjects, z-scores, and the loading vector are shown for the first principal component for the hip flexion moment. Extreme subjects are moments characterizing high and low z-scores. Subjects with high z-scores express more of the trend depicted by the loading vector.

Waveform features characterizing model-independent differences between control and osteoarthritis groups were assessed using loading vectors and z-scores. Extreme subjects are shown for interpretation in Figure 3.6. Regardless of the joint model, subjects with osteoarthritis exhibited significantly decreased hip adduction magnitude and increased knee adduction magnitude. Hip adduction moments were also flatter in mid-stance for subjects with osteoarthritis. Changes in amplitude for knee flexion and ankle plantarflexion moments were not statistically significant. There was no waveform difference between groups for hip internal rotation and ankle eversion moments.
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Figure 3.6: Four model-independent changes due to osteoarthritis
Extreme subjects are shown for control (solid black line) and osteoarthritis (dashed grey line) groups to characterize high and low principal component z-scores. Arrows give the direction of the osteoarthritis group relative to the control group throughout the shaded regions. Principal component analysis identified magnitude differences for the hip and knee adduction moments, and amplitude differences for the knee flexion and ankle plantarflexion moments.

3.4 Discussion

This study is consistent with previous findings of significant changes due to medial compartment knee osteoarthritis in lower limb joint moments in the frontal, sagittal, and transverse planes [15,48,63]. We hypothesized that while group differences in some joint moments would be affected by the choice of joint model, other measures would represent changes due to osteoarthritis across all joint models. The results of this study support our hypothesis, as tested using both discrete peak parameters and principal component analysis.
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Of the nine locations where significant group differences were detected in discrete peak parameters, five depended on the choice of joint model (Table 3.1). Such model-dependency raises the question of whether these differences are truly due to knee osteoarthritis, or merely artifacts of inaccurate joint models or subjective waveform parameterization. Principal component analysis was used to extract waveform features that integrate the model-dependent discrete peak parameters with overall changes in the waveforms. The hip flexion moment was the only case where the principal component capturing a group difference was dependent on the choice of joint model.

There was a significant group difference at the late-stance peak when all hip flexion moments were expressed in the PoP model along a common axis perpendicular to the plane of progression. During stance, pelvic rotation is minimal [64], therefore the proximal flexion axis, shared with the Joint Coordinate System, does not deviate excessively from PoP flexion axis. However, incorporating a subject-specific flexion axis in the Proximal model further increased the significance of the group difference. This late-stance difference corresponds with previous findings [15,48,63]. However, there was no evidence of any group difference in the Distal model, where the flexion axis follows the internal rotation of the femur. Furthermore, PC1, describing this late-stance peak difference as part of an overall increase in magnitude for subjects with osteoarthritis, was significant only in the Proximal model. For hip flexion moments, the choice of joint model affects the detection of group difference due to osteoarthritis using either discrete peak parameters or principal component analysis.
In this study, subjects with osteoarthritis exhibited reduced hip adduction moments during stance in all joint models, as measured by PC1 and the second peak. This overall reduction corresponds with findings by Mundermann et al. [15] using a PoP joint model for subjects with severe (K/L grade $\geq 3$) osteoarthritis. In contrast, Huang et al. [63] found an increase in hip adduction moment at the second peak for subjects with severe osteoarthritis, which was attributed to an attempted protective mechanism of pelvic list to reduce the knee adduction moment. While Huang et al. [63] did not provide enough information to identify their choice of joint model, our results suggest that this discrepancy is not a product of differences in joint models between studies. It is important to note that the mean walking speed of all subjects in the study by Huang et al. [63] was approximately 30% slower (0.89 vs 1.30 m/s) than that of the subjects in both the current study and the study by Mundermann et al. [15]. This appears to be a substantial difference, given that walking speed has been shown to influence other lower-limb joint moments [65,66]. In all models, the osteoarthritis group also demonstrated a reduced unloading of the hip adduction moment during mid-stance, as evidenced by the increase in the mid-stance to peak ratio given by PC3.

Increased knee adduction moment magnitude throughout stance, represented by PC1, was a robust characteristic of the osteoarthritis group across all joint models in accordance with previous studies that used principal component analysis [44,54,67]. The mid-stance minimum value was also greater for the osteoarthritis group in all joint models as noted by Weidenhielm et al. [68]. However, in contrast to findings by Newell et al. [54], neither the first nor the second peak was significantly different between groups.
in any joint model. This is not the first investigation to find no significant difference in peak adduction moments due to osteoarthritis [56]. In this study, peak values were taken from a window around the mean peak location, which may explain the failure to find significant peak differences as identified in previous studies at specific locations [54] and across the entire stance phase [46,58,69,70]. Such contradiction in the literature may be due to the inconsistency and subjectivity of discrete peak parameters as applied to pathological gait. Our results support the conclusion of Newell et al. [54] that changes in the overall magnitude given by PC1, as well as the mid-stance minimum value, are characteristics of osteoarthritis that are not affected by the choice of joint model.

Some differences in joint moments were clearly artifacts of joint model selection. Early stance transverse plane hip moments in the Proximal model were internal rotation moments, while the other three models exhibited external rotation moments as found by Schache et al. [13]. Taking a sagittal view, this difference corresponds with the hip flexion angle, which starts around 30 degrees and declines toward zero after mid-stance [64]. In early stance, the Proximal model thus has an internal rotation axis partially aligned with both the Distal model’s adduction and internal rotation axes, and its positive magnitude is dominated by a contribution from what in the Distal model would be called an adduction moment. Such cross-talk is also evident for knee internal rotation and ankle eversion moments, which are similarly influenced by large sagittal plane joint angles between the long axes of proximal and distal segments.
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Hip internal rotation moment differences between osteoarthritis and control groups were found only in the PoP model at the early-stance peak. For the hip internal rotation moment, an argument could be made that the difference found only in the PoP model has no physiological significance. An internal rotation axis projected onto the plane of progression does not follow the long axis of the limb. However, Astephen et al. [48] identified this reduction in early-stance peak moment as a progressive characteristic of knee osteoarthritis using the JCS model and McKean et al. [71] found a significant principal component describing a decreased amplitude effect. The failure to find a significant PC1 amplitude effect in this study may reflect the disease severity of this particular group of osteoarthritis subjects.

In the Distal and Proximal models, subjects with osteoarthritis exhibited a reduction in the early-stance knee external rotation (negative internal rotation) moment as found in previous studies using both principal component analysis and discrete peak parameters [57,71]. In contrast, in the PoP model the osteoarthritis group had an increased positive internal rotation peak in early stance. Regardless of the presence of cross-talk between joint models, there is a significant change in early-stance knee internal moments due to knee osteoarthritis. However, the physiological meaning of this change is model-dependent. The reduced late stance knee internal rotation peak, found only in the Proximal model, has not been previously reported as a characteristic of subjects with osteoarthritis. In this study, the presence of reduced magnitudes in both early and late stance led to a principal component that captured an overall amplitude reduction across all models as a characteristic of subjects with osteoarthritis.
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We have shown that joint moments are not merely affected by the choice of joint model; joint moments are affected differently for subjects with medial knee osteoarthritis as compared with asymptomatic controls. Both the detection and interpretation of group differences are confounded by the use of varying joint models in literature.

If the goal of moment analysis is to relate joint loads to joint rotations, and therefore muscle activations, then joint moments should be reported in a model corresponding to joint rotations [13,72]. According to ISB recommendations this is the Joint Coordinate System [22,34]. We found only two cases where the Joint Coordinate System could not be used to detect a difference in joint moments between subjects with and without osteoarthritis: early-stance hip internal rotation and early-stance knee flexion. However, the discrete peak parameters in both cases were different only in the PoP model, which, being planar, has the least anatomical relevance. Expressing moments in the Joint Coordinate System did not mask meaningful changes due to osteoarthritis. Our results support the adoption of the Joint Coordinate System as a standard for the expression of joint moments in studies of osteoarthritis.

It is important to note that joint moments may also be used as surrogates for joint contact forces. The knee adduction moment has been shown to correlate with the mediolateral distribution of knee joint contact forces [14]. However, this correlation has only been demonstrated using the Distal model, and may not be accurate in the Proximal model where the adduction axis deviates from the tibial plateau during flexion [23]. This issue need not preclude the standard adoption of the Joint Coordinate System for
CHAPTER 3: THE EFFECT OF JOINT MODELS

reporting joint moments as its adduction axis does not deviate from the tibial plateau during flexion. Future studies could evaluate the alternative joint models in their capacity to predict joint contact forces.
Chapter 4

Conclusion and Recommendations

There is no debate in the literature [13,24,27,73], or in the results presented in Chapter 3, regarding the fact that the choice of joint model has a significant impact on the shape and magnitude of the joint moment waveform during gait. In particular, transverse plane joint moments have been shown to be particularly sensitive to the choice of joint model [73]. In order to select an appropriate joint model it is important to understand the causes of the inter-model differences.

4.1 The Mechanism of Model-dependency

Schache et al [73] presented a thorough investigation of the mechanism leading to such differences in the transverse plane moments. They first noted that the ground reaction force was the dominant contributor to transverse plane moments during stance. Next, they presented a projection of the ground reaction force onto the transverse planes of the global laboratory and anatomical transverse planes for the ankle, knee, and hip joints. Comparing the coordinate systems in Figure 4.1 it is clear that the magnitude and
direction of the projected ground reaction force with respect to the anatomical axes depends on the chosen transverse plane.

![Figure 4.1: Mechanism of model-dependency in transverse plane moments](image)

Orientation of the resultant ground reaction force (GRF) vector projected into (a) the laboratory frame (LF) transverse plane, (b) the femoral anatomical frame (AF) transverse plane and (c) the tibial (proximal) AF transverse plane during initial stance (left column) and terminal stance (right column) for the left lower limb. The projected GRF vector is indicated by the thick solid line. Note that LF (HJC) and LF (KJC) represent the translated LF such that the origin is located at the HJC and KJC, respectively [73].

Newell [74] presented a similar discussion for the frontal plane adduction moment at the knee. Projecting the net external knee moment vector and the corresponding knee adduction axes for three joint models onto the global transverse plane, it was shown in Figure 4.2 that there is a significant difference in the relative orientations between the...
three models and the net moment vector. Thus, the differences in joint moments between anatomical joint models are simply due to differences in orientation between the joint models.

Figure 4.2: Three alternative knee adduction axes vs. net moment
Net reaction moment (black) and adduction axes for three models (yellow=PoP, orange=Distal, green=JCS) plotted in the global transverse plane for the asymptomatic group at 25% (a), 50% (b) and 75% (c) of the stance cycle. The global negative y-direction corresponds to the direction the subjects walked, the global x-direction points in the lateral direction of the subject’s leg [74].

4.2 The Mechanism of Group-model Interaction

The fact that the expression of joint moments is so closely linked to joint kinematics bears consideration. It is well documented that subjects with gait pathologies such as osteoarthritis exhibit joint kinematics that are distinctly different from asymptomatic control subjects [48,75]. It is also certainly understood that different forces will be developed by osteoarthritis subjects to obtain the altered kinematics. However, it seems less obvious that even if the forces were the same, inter-group differences in joint kinematics can produce the appearance of differing forces and moments. For example
CHAPTER 4: CONCLUSION AND RECOMMENDATIONS

consider a situation where a subject rotates his pelvis forward 90 degrees such that the torso is parallel to the ground, yet the subject maintains a typical ground reaction force and moment profile. If the joint model used to report hip moments was fixed in the proximal (upper) segment, then the internal rotation axis for this subject would coincide with the adduction axis of an erect subject, and the leaning subject’s joint moments for these two axes would be reversed. Using the Proximal joint model, it would be misleading to compare the moments of this leaning subject with those of erect control subjects. However, using the Distal joint model, the axes for the two groups might be comparable. This is an extreme example, but it illustrates how interpretation of components of the net moment at a joint can be confounded by group differences in the orientation of the reference frames. In this study, as the net joint moments were constant across all joint models, the differences in the way each group was affected by the choice of joint model can only be attributed to group differences in joint kinematics.

4.3 Considerations in Joint Model Selection

In Chapter 3, it was concluded that the Joint Coordinate System provides the greatest clinical relevance for joint moments due to its relationship to joint rotation axes, and therefore to muscle activation. There are other factors that could be important in joint model selection, as illustrated in Figure 4.3.
What is the goal of moment analysis?

Anatomical Differences

Joint Coordinate System

Other anatomical model

Non-Anatomical Differences

Global Reference Frame

Figure 4.3: Joint model selection

Studies based on disease characterization may support a reference frame that is common for all subjects: a global reference frame with a fixed orientation relative to the plane of progression. Alternatively, studies interested in the physiological impact of changes in moments require an anatomically based coordinate system.

4.3.1 Non-anatomical differences

If the primary objective of a study is simply to find parameters that correlate with the incidence and progression of osteoarthritis, there may be no need to introduce an anatomical coordinate system. A comparison of joint moments between groups in a common reference frame, such as the global laboratory system, avoids the issue of uncertainty in the orientation of anatomical axes. Changes in the magnitude and orientation of the net joint moment expressed in a laboratory-fixed reference frame could reveal different gait strategies for subjects with osteoarthritis that are not artifacts of the
choice of joint model. An example of this approach is the hip flexion moment when calculated using the PoP model in this study. Comparing subjects using a common laterally-directed flexion axis, a significant difference was found in late-stance hip flexion as shown in Table 3.2 and Figure 3.2. It is also possible that there might exist another laboratory-fixed axis that would reveal an even greater group separation along an arbitrary direction. Such parameters are appealing because of their relative simplicity, repeatability, and ease of measurement. In this particular case, the PoP flexion axis was closely aligned with the anatomical flexion axis; therefore, it was also associated with some clinical significance.

However, in the same way that incorporating subject-specific inertial parameters reduces kinematic variability, it seems reasonable to expect that subject-specific anatomical alignment of the coordinate system will reduce inter-subject variability in measured joint moments. Indeed, expressing the hip flexion moment in the Proximal anatomic joint model revealed a greater separation between osteoarthritis and control groups than when using the common PoP flexion axis (Table 3.2). Furthermore, as the anatomical axes are required to calculate the kinematics, and therefore joint moments, there is not a great reduction in the model complexity or computational cost by analyzing moments in a fixed reference frame. Lastly, even if moments are not different between groups in the fixed global reference frame, differing limb orientations captured by anatomical joint models will affect the muscular response and contact force distribution. Both of these are parameters with direct relevance to the pathomechanics of knee osteoarthritis.
CHAPTER 4: CONCLUSION AND RECOMMENDATIONS

4.3.2 Anatomical Differences

In most studies, therefore, it would be preferable to analyze moments using a subject-specific model based on anatomical features. Selecting an anatomical joint model for the expression of joint moments revolves around two questions:

1. Does the chosen joint model increase or decrease the variability in the measured joint moments?
2. Does the chosen joint model have a physiological interpretation?

The first question could be rephrased to ask whether the joint model increases or decreases the ability to detect a difference between groups with and without osteoarthritis. For example, the hip flexion moment, when expressed in the Distal joint model, exhibited a much larger standard deviation throughout the stance phase than when expressed in the other three models (Figure 3.2). Indeed, the late-stance difference between osteoarthritis and control groups in the PoP, JCS, and Proximal models disappeared when the hip flexion axis was defined in the Distal joint model (Table 3.2).

Kadaba et al. [64] report hip internal rotation angles during stance in the range of 5-35 degrees and hip adduction angles between -5 and 10 degrees for asymptomatic control subjects, which would seem to implicate joint kinematics as a source of variability. Furthermore, Briem et al. [76] identified a reduction in the stance-phase hip adduction angle for affected versus unaffected limbs in subjects with unilateral knee osteoarthritis. Thus, due to kinematic variability, the Distal model was unable to detect a difference in hip flexion moments between groups.
CHAPTER 4: CONCLUSION AND RECOMMENDATIONS

Increased variability may not, alone, be grounds for dismissing the Distal joint model for expression of hip flexion moments if it could be argued that this model best represents anatomical differences. However, for lower-limb flexion moments the primary flexion rotation occurs about an axis that is approximately fixed in the proximal segment, not the distal segment [22,77]. Flexion moments reported about a distal segment axis have no clinical meaning. Thus, expressing moments along an unsuitable anatomical axis can actually inhibit the detection of differences due to osteoarthritis. While the confounding influence of the Distal model was less evident for knee and ankle flexion, the same argument of anatomical relevance applies in terms of the functional axes of these joints. A similar argument was developed in Chapter 3 regarding the anatomical relevance of internal rotation moments reported in the proximal segment.

For some specific parameters, the choice of model may be less important. The mid-stance adduction moment has been shown to exhibit group differences due to osteoarthritis regardless of the choice of joint model [24]. Referring to Figure 4.2 b), at mid-stance the alternative knee adduction axes are closely aligned, therefore the choice of model has less effect on the mid-stance moment magnitude. A description of all model-independent moment parameters detected in this study was provided in Chapter 3, and is summarized in Section 4.7. In studies that employed a joint model other than the Joint Coordinate System, researchers were cognizant of the fact that their joint model may only be valid for specific moment parameters. Mundermann et al. [15] did not report internal rotation moments, perhaps because their PoP internal rotation moment axis was not aligned with the functional internal rotation axis fixed in the distal segment.
Furthermore, no studies were found to report hip flexion moment using the distal joint model. Nevertheless, while the usage of alternative joint models may be valid for specific parameters, it limits the potential for analysis in all anatomical directions, and it introduces unnecessary confusion into the study of joint moments.

4.3.3 Joint Coordinate System

The argument in favour of adopting the Joint Coordinate System as a standard for the expression of lower-limb joint moments is based on relatively recent publications that proposed to standardize the lower-limb model of joint kinematics [22,34]. It has been noted, both in this study and in others, that joint moments should be reported in directions corresponding to joint rotations [72,78]. However, when the Joint Coordinate System was proposed as a standard for joint rotations, the authors specifically cautioned against its use for reporting joint moments because of its non-orthogonality [22]. The aim in the standardization of the reporting of kinematic data was to preserve the connection between biomechanical data and clinical medicine. This same goal should apply to the expression of joint moments and forces. If the moments are intended to be used as outcome measures, there is no issue with a non-orthogonal joint model as long as it is recognized as such. However, if joint moments components will later be recombed in a calculation of joint power they should be calculated using an orthogonal coordinate system. This is an important distinction that must not be ignored, but that will allow the reported joint moments to correspond with clinically meaning directions.
CHAPTER 4: CONCLUSION AND RECOMMENDATIONS

4.4 Conclusion

It was hypothesized that the choice of joint model would affect the detection of differences in lower-limb joint moments between subjects with and without knee osteoarthritis. Comparing forty-four symptomatic subjects, and forty-four asymptomatic controls, it was found that the choice of joint model significantly affected the detection of differences in joint moments. This model-dependency was attributed to altered kinematics in the osteoarthritis group.

It was also hypothesized that some features of the joint moment waveforms would exhibit group differences regardless of the choice of joint model. The magnitude and timing of discrete peak parameters taken from the waveforms were significantly different between joint models. Similarly, the principal component z-scores were significantly different between joint models. Thus, when lower-limb joint moments are calculated using two different joint models the absolute magnitudes of waveform parameters cannot be directly compared. However, some features of joint moment waveforms were found to change in a consistent, if not equal, manner across all joint models in the presence of osteoarthritis. Significant model-independent differences were found in hip adduction moments, knee adduction moments, knee internal rotation moments, and ankle plantarflexion moments. These changes can be attributed solely to the presence of moderate knee osteoarthritis.

The final objective was to identify the joint model that was most suitable for detecting clinically meaningful changes in joint moments due to osteoarthritis.
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Transverse plane joint moments were found to be extremely sensitive to the choice of joint model, and their interpretation could vary from internal rotation to external rotation across models. However, it was proposed that in agreement with the goal of attaching clinical meaning to the reported moments, the internal rotation axes should be aligned with the long axis of the distal segment; this is the same axis used to report internal rotation angles. Thus, the PoP and Proximal joint models were not suitable for the expression of internal rotation moments, and these parameters should be analyzed using the Distal or Joint Coordinate System models, which share the distal internal rotation axis. Similarly, the Distal model was found to be unsuitable for the expression of flexion moments.

This study was unable to conclude that any of the alternative anatomic joint models are mathematically incorrect, or to quantify one alternative as being superior to the others. However, the heuristic argument in favour of the Joint Coordinate System was supported by the fact that there were only two cases where it could not be used to detect differences in joint moments between subjects with and without medial knee osteoarthritis: early-stance hip internal rotation and knee flexion. Furthermore, in both of these cases the difference was only detected using the PoP model, which does not correspond precisely with any anatomically or clinically meaningful directions.
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4.5 Implications for Knee Osteoarthritis

This thesis provides a framework for comparing the joint moments reported by research groups using alternative joint models. Specifically, features of joint moment waveforms were identified that are characteristic of medial knee osteoarthritis regardless of the choice of joint model. These features are valuable because their reliability as outcome measures facilitates the development and comparison of clinical interventions. Attention should be directed toward changes in hip adduction moments, knee adduction moments, knee internal rotation moments, and ankle plantarflexion moments. In addition, future studies should not ignore changes in hip internal rotation moments and knee flexion moments that in this study verged on significance, and have been shown to progress with disease severity [79].

Features were also identified that can separate groups with and without osteoarthritis only when an appropriate joint model is used. The hip flexion moment should only be reported using the Proximal joint model axis (shared with the JCS), and all internal rotation moments should only be reported using the Distal joint model axis (shared with the JCS). Results supported the adoption of the Joint Coordinate System as a standard joint model for expression of lower-limb joint moments in studies of knee osteoarthritis.

Knee adduction moment peaks were not significantly different between subjects with osteoarthritis and asymptomatic control subjects, regardless of the joint model. This is contrary to many previous studies [46,58,69,70]. In this thesis, the sample size of 44 subjects per group provided the statistical power to detect a 13% change in moment peaks.
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at a significance level of $\alpha = 0.05$. This change is comparable to effect sizes reported to be clinically relevant [24,47,80], yet larger than the 10% observed difference.

The conclusion of this thesis is not that a change cannot be found in knee adduction moment peaks due to knee osteoarthritis; indeed, a different subject group and laboratory conditions could well provide the additional 3% change in moment peaks required to obtain statistical significance. However, the issue is that adduction moment peaks cannot reliably be used to indicate the presence of moderate radiographic osteoarthritis because results depend both on the subject group and on the choice of joint model. As outlined above, this study has identified alternative measures that may be superior for use in characterizing medial knee osteoarthritis and in developing clinical interventions to mitigate the effects of mechanical loading in osteoarthritis progression. For the knee adduction moment, these parameters are the mid-stance magnitude and an overall measure of magnitude provided by principal component analysis.

It is not known whether there is some critical threshold for loading which results in knee osteoarthritis, or whether the disease might result from cumulative loading or subtle changes in loading strategies. For this reason, both discrete measures of moment magnitude and principal component analysis provide valuable data. In this study, principal component analysis yielded waveform differences that were less sensitive to the choice of joint model than discrete peak parameters. However, if standardization of the joint model for reporting lower-limb joint moments is achieved, peak parameters will be much more comparable between studies, and may still be worth investigation.
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This study has identified model-independent parameters that can be used to assess the effectiveness of clinical interventions aimed at restoring normal joint kinetics to subjects with medial knee osteoarthritis.

4.6 Limitations

The primary limitation in this study was the subject groups. Significant differences were detected in height, weight, age, and body mass index. Subject groups were also imbalanced in terms of gender. These changes could have introduced error to inter-group comparisons that was not due to the presence of osteoarthritis. Subjects in the osteoarthritis group were not stratified according to radiographic osteoarthritis severity as in some studies [15,63]. However, separation into moderate and severe osteoarthritis groups based on KL scores is not standardized [15,63], and the definition of moderate osteoarthritis in this study corresponds with other investigations [24,48]. The joint model-dependency observed in this study cannot be extrapolated to subjects with severe osteoarthritis, although similar affects would be expected due to their kinematic differences.

A second, and perhaps no less important, limitation of this study is the error inherent in gait analysis using the current techniques. Error in the definition of joint centres, joint axes, inertial parameters, and skin motion may have inhibited the detection of differences due to osteoarthritis. A further limitation is the anatomical relevance of the joint axes for the ankle. It is known that the ankle joint involves rotations between the shank and the talus, and between the talus and the foot; these two joints are commonly called the
talocrural, and subtalar joints, respectively [36]. While the talocrural flexion axis can be approximated by the inter-malleolar axis as in this study, moments about the adduction and eversion axes may not have direct anatomical interpretations. All of these modeling assumptions hinder the comparison of these results with other investigations.

Lastly, this study does not identify the specific kinematic differences that resulted in group-model interactions for joint moment parameters. The aim of this study was not to identify how kinematic differences affect joint moments, but rather to identify how joint moments are affected by kinematic changes in the joint model used for interpretation. Kinematic differences can easily be quantified and analyzed independently of joint kinetics. However, identification of the precise influence of joint kinematics in confounding joint moment interpretation could lead to the development of superior coordinate system definitions.
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4.7 Recommendations

The recommendations outlined in this thesis are as follows:

1. Adopt the Joint Coordinate System as a standard joint model for the expression of lower-limb joint moments in studies of knee osteoarthritis, and perhaps more generally for all gait analysis.

2. Direct attention toward model-independent joint moment parameters for characterization of knee osteoarthritis. These include:
   i. Late-stance hip adduction
   ii. Mid-stance knee adduction
   iii. Early-stance knee internal rotation
   iv. Late-stance ankle plantarflexion
   v. Overall hip adduction magnitude
   vi. Overall knee adduction magnitude
   vii. Overall knee internal rotation moment amplitude

3. Employ principal component analysis to detect joint model-insensitive features of lower-limb joint moments.
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4.7.1 Future work

Before the Joint Coordinate System can be adopted as a standard for the expression of joint moments, it must be confirmed that the knee adduction moment calculated using this joint model is well correlated with the mediolateral distribution of joint contact force on the tibial plateau. This relationship has been developed for the Distal model, and it is speculated that the Proximal model is unsuitable [14,23]. As this promises to be an important parameter for the pathomechanics of knee osteoarthritis, the Joint Coordinate System must prove equally suitable.

Furthermore, the effect of joint models on the detection of group differences in joint moments for other gait pathologies could influence the widespread adoption of a standard biomechanical model for interpretation.

Finally, the argument for standardized reporting of joint moments in the Joint Coordinate System was based on its anatomical relevance. However, the anatomical axes used in this study for the hip, knee, and ankle are approximations that do not reflect the true kinematics of each joint. Research should continue to be directed toward superior description of joint kinematics, and ultimately may lead to the adoption of a new standard for the expression of joint rotations. In such event, it may be appropriate to adopt a new method of interpreting joint moments for enhanced clinical meaning.
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