

THE BIOMECHANICAL EFFECTS OF AN EXTERNALLY APPLIED
ORTHOSIS ON MEDIAL COMPARTMENT KNEE OSTEOARTHRITIS

by

Janet Conrad

Submitted in partial fulfillment of the requirements
for the degree of Master of Applied Science

at

Dalhousie University
Halifax, Nova Scotia
December 2011

© Copyright by Janet Conrad, 2011

DALHOUSIE UNIVERSITY

DEPARTMENT OF BIOMEDICAL ENGINEERING

The undersigned hereby certify that they have read and recommend to the Faculty of Graduate Studies for acceptance a thesis entitled “THE BIOMECHANICAL EFFECTS OF AN EXTERNALLY APPLIED ORTHOSIS ON MEDIAL COMPARTMENT KNEE OSTEOARTHRITIS” by Janet Conrad in partial fulfillment of the requirements for the degree of Master of Applied Science.

Dated: December 6, 2011

External Examiner:

Co-Supervisors:

Readers:

DALHOUSIE UNIVERSITY

DATE: December 6, 2011

AUTHOR: Janet Conrad

TITLE: THE BIOMECHANICAL EFFECTS OF AN EXTERNALLY APPLIED
ORTHOSIS ON MEDIAL COMPARTMENT KNEE OSTEOARTHRITIS

DEPARTMENT OR SCHOOL: Department of Biomedical Engineering

DEGREE: M.A.Sc.

CONVOCATION: May

YEAR: 2012

Permission is herewith granted to Dalhousie University to circulate and to have copied for non-commercial purposes, at its discretion, the above title upon the request of individuals or institutions. I understand that my thesis will be electronically available to the public.

The author reserves other publication rights, and neither the thesis nor extensive extracts from it may be printed or otherwise reproduced without the author's written permission.

The author attests that permission has been obtained for the use of any copyrighted material appearing in the thesis (other than brief excerpts requiring only proper acknowledgement in scholarly writing), and that all such use is clearly acknowledged.

Signature of Author

TABLE OF CONTENTS

List of Tables	vii
List of Figures	viii
Abstract	x
List of Abbreviations and Symbols Used	xi
Acknowledgements	xiii
Chapter 1 Introduction	1
Chapter 2 Background	4
2.1 Knee Osteoarthritis	4
2.1.1 Prevalence and Cost	4
2.1.2 Illness and Disease	5
2.1.3 Biomechanical Changes with Knee OA	6
2.2 Gait Analysis and Knee OA	6
2.2.1 Kinematics	7
2.2.2 Kinetics	9
2.2.3 Neuromuscular Activation	12
2.2.4 Alterations in Medial Compartment Loading with Knee OA During Gait	13
2.2.5 Alterations in 3D Kinetics and Kinematics with Knee OA During Gait	13
2.2.6 Principal Component Analysis	15
2.3 Treatments	17
2.4 Bracing	18
2.4.1 Bracing as a Treatment of Illness	18
2.4.2 Bracing as a Treatment of Disease	19

Chapter 3	Does the Valgus Unloader Brace Affect the External Knee Ad- duction Moment Magnitude?	26
3.1	Introduction	26
3.2	Methods	29
3.2.1	Participants	29
3.2.2	Gait Analysis	30
3.2.3	Data Processing	31
3.2.4	Principal Component Analysis (PCA)	31
3.2.5	Brace vs. No Brace Analysis: Knee Adduction Moment Magnitude	32
3.3	Results	32
3.3.1	Brace vs. No Brace Analysis: Knee Adduction Moment Magnitude	32
3.4	Discussion	35
3.4.1	Conclusion	38
Chapter 4	Does the Brace Affect All Participants in the Same Way?	39
4.1	Introduction	39
4.2	Methods	43
4.2.1	Data Processing	43
4.2.2	Principal Component Analysis (PCA)	43
4.2.3	Analysis of Groups	44
4.2.4	Varus Thrust	45
4.2.5	Statistical Analysis	45
4.3	Results	47
4.3.1	Participants	47
4.3.2	Two-Way ANOVA	48
4.4	Discussion	57
4.4.1	Conclusion	61
Chapter 5	Conclusion	62
5.1	Summary of Thesis Objectives	62
5.2	Summary of Chapter 3: Does the Valgus Unloader Brace Affect the Exter- nal Knee Adduction Moment Magnitude?	63

5.3	Summary of Chapter 4: Does the Brace Affect All Participants in the Same Way?	63
5.4	Clinical Implications	65
5.5	Limitations	67
5.6	Future Research	68
5.7	Conclusion	70
Appendix A	Loading Vectors	71
Appendix B	Varus Thrust Waveforms of Participants Removed from the Analysis	90
Bibliography	92

LIST OF TABLES

3.1	Participant demographics	35
4.1	Group demographics	47
4.2	Results of two-way ANOVA	50
4.3	Post-hoc analysis of condition effects	51
4.4	Varus thrust magnitude, maxima, and minima by condition	51

LIST OF FIGURES

2.1	IREM marker set-up	8
2.2	Joint coordinate system axes	9
2.3	2D inverse dynamics	11
2.4	An example of Principal Component Analysis	16
2.5	Valgus unloader brace (Breg brand)	19
2.6	Distribution problem with brace	22
3.1	Valgus unloader brace (Breg brand)	29
3.2	Knee adduction moment PC1	33
3.3	Knee adduction moment by condition	34
3.4	Distribution problem with brace	37
4.1	Example of varus thrust calculation	46
4.2	Knee adduction moments by group	49
4.3	Condition effects	52
4.4	Group trend	53
4.5	Knee rotation moment magnitude interaction	54
4.6	Knee flexion angles magnitude interaction	55
4.7	Knee rotation angle PC2 interaction	56
A.1	Knee adduction moment PC1	72
A.2	Knee adduction moment PC2	73
A.3	Knee adduction moment PC3	74
A.4	Knee flexion moment PC1	75
A.5	Knee flexion moment PC2	76
A.6	Knee flexion moment PC3	77
A.7	Knee rotation moment PC1	78
A.8	Knee rotation moment PC2	79

A.9	Knee rotation moment PC3	80
A.10	Knee adduction angle PC1	81
A.11	Knee adduction angle PC2	82
A.12	Knee adduction angle PC3	83
A.13	Knee flexion angle PC1	84
A.14	Knee flexion angle PC2	85
A.15	Knee flexion angle PC3	86
A.16	Knee rotation angle PC1	87
A.17	Knee rotation angle PC2	88
A.18	Knee rotation angle PC3	89
B.1	Knee adduction angle waveform of participant in the Decrease group removed from varus thrust analysis	91
B.2	Knee adduction angle waveform of participant in the Increase group removed from varus thrust analysis	91

ABSTRACT

This thesis examined the immediate biomechanical effects of valgus unloader brace application in participants with moderate medial compartment knee osteoarthritis during gait. Thirty-three individuals were prescribed a valgus unloader brace. 3D knee moments and angles were calculated during walking with and without the brace. Principal Component Analysis identified amplitude and temporal changes of the moment and angle waveforms during gait. Three groups were identified based on the change in knee adduction moment magnitude with brace application. Two-Way ANOVA tested for differences among groups and conditions in principal component scores, as well as discrete varus thrust values.

There existed three subgroups of participants identified by different gait adaptations to brace application. The brace had temporal and magnitude effects on 3D kinetics and kinematics for the participant group. This study showed that the brace does not provide a consistent change to knee joint mechanics. These results have implications for brace prescription.

LIST OF ABBREVIATIONS AND SYMBOLS USED

Abbreviation	Long Form
OA	Osteoarthritis
PCA	Principal Component Analysis
PC	principal component
WOMAC	Western Ontario and McMaster Universities Osteoarthritis Index
3D	Three dimensional

Symbol	Description	Units
$\sum F$	sum of the forces	N
m	mass of limb segment	kg
a	linear acceleration of limb segment	m/s ²
F _{dist}	force of the distal limb segment acting on the limb	N
F _{prox}	force of the proximal limb segment acting on the limb	N
g	acceleration due to gravity	m/s ²
$\sum M$	sum of the moments	N • m
I	moment of inertia of the limb	kg • m ²
α	angular acceleration of limb segment	rad/s ²
M _{dist}	moment of the distal limb segment acting on the limb	N • m
M _{prox}	moment of the proximal limb segment acting on the limb	N • m
r _{dist}	moment arm between the line of action of F _{dist} and the limb center of mass	m
r _{prox}	moment arm between the line of action of F _{prox} and the limb center of mass	m
N	number of participants	
X	matrix of all participant waveforms for each waveform type (101 X N)	

Symbol	Description	Units
nPC	number of extracted principal components for each waveform type	
\mathbf{U}	matrix of extracted eigenvectors for each waveform type	
f_{muscle}	force in muscle	N
f_{ligament}	force in ligament	N
f_{brace}	force in brace	N
F_{net}	net external force at knee	N
M_{net}	net external moment at knee	N • m
$\Delta\%PC$	percent change in PC score	
PC_{brace}	PC score during the brace condition	
$PC_{\text{no brace}}$	PC score during the no brace condition	
x_t	$\Delta\%PC$ threshold value used for subgroups	
$\Delta\%PC_{\text{max}(+)}$	greatest positive percent change in PC score over all participants	
$\Delta\%PC_{\text{max}(-)}$	greatest negative percent change in PC score over all participants	

ACKNOWLEDGEMENTS

Throughout my work and study in the Dynamics of Human Motion Laboratory, I have always been able to find help when I needed it. I have been lucky to work with such a dynamic and diverse group of researchers and would like to thank those who have always been there to answer my questions: Gillian Murdock, who knows every reference; Heather Butler, the Minitab Queen; Derek Rutherford, who is always up for talking data; Graeme Harding, for filling my whiteboard with flow charts and understanding what it's like to wait; Shawn Robbins, who is always looking out for my best interests; and Nick Hill, who has always answered all my questions, relevant or otherwise.

I would also like to thank the funding bodies that have placed their trust in my research abilities: the Natural Sciences and Engineering Research Council of Canada, the Nova Scotia Health Research Foundation, and particularly the Killam Trusts. Being a part of the Killam legacy, a family that has made such a large impact on this province, has been a true highlight of my Dalhousie experience.

I have been very fortunate to work with a supervisory committee set on fostering an environment in which I was able to learn from my mistakes and take pride in my accomplishments. I would like to thank the members of my supervisory committee. Dr. William Stanish has always been able to put my research into the broader, clinical perspective even at the times when it felt far removed. I have been fortunate to learn from Dr. Graham Caldwell, whose insightful skill at putting complex subjects into simple terms was greatly appreciated. And finally, I would like to thank my supervisors, Dr. Janie Astephen Wilson, who has shown me the importance of being true to the data and Dr. Cheryl Kozey, who provided me with my first opportunity in Biomedical Engineering when she hired me as a co-op student. It was hard to leave once I'd had the ice cream cake.

CHAPTER 1

INTRODUCTION

Osteoarthritis (OA) is the most prevalent type of arthritis in Canada (The Arthritis Society of Canada, 2004). It is a leading cause of disability among adults worldwide (Neogi et al., 2011) and the pain and limited mobility associated with the disease can lead to a lower quality of life (Neogi et al., 2011). OA affects 1/10th of Canadians and similar numbers are seen in the USA - estimates of the prevalence of knee OA are between 19-28% of adults over 45 years old (Felson et al., 1987; Jordan et al., 2007). Increasing numbers of younger adults between the ages of 40 and 55 receiving joint replacements (Canadian Institute for Health Information, 2009; Jain et al., 2005) highlights the need for investigation of non-surgical interventions that will slow the progression of knee OA.

Although the disease has both a biomechanical and biochemical etiology, it is thought to be mechanically induced (Brandt et al., 2006). In the knee, OA occurs more commonly in the medial compartment than the lateral (Felson and Radin, 1994). The external knee adduction moment magnitude, measured using gait analysis, has been associated with medial compartment contact forces in the knee and is therefore often considered a surrogate measure of medial compartment load (Erhart et al., 2010; Zhao et al., 2007). Both severity and progression of medial compartment knee osteoarthritis have been linked to changes in the knee adduction moment waveform. High peak external knee adduction moments have been associated with knee OA progression and severity (Andriacchi and Mündermann, 2006; Mündermann et al., 2005; Miyazaki et al., 2002) and changes in the overall waveform shape over stance phase have been associated with severity (Astefhen et al., 2008). Therefore, increased medial compartment contact forces are related to medial

compartment knee OA progression and high knee adduction moments are an indicator of patients' risk of disease progression (Andriacchi and Mündermann, 2006).

There is an increasing demand for conservative treatments of OA (Buckwalter et al., 2001). Valgus unloader braces are a type of non-operative treatment that are sometimes prescribed to patients with mild to moderate medial compartment knee osteoarthritis to relieve pain and increase stability (Zhang et al., 2008). They have been shown to improve self-reported pain and function (Gaasbeek et al., 2007; Richards et al., 2005; Kirkley et al., 1999). Valgus unloader braces are purported to apply an external abduction moment at the knee to reduce the medial compartment load associated with medial compartment knee OA progression (Pollo et al., 2002).

The effects of valgus unloader bracing on the net external knee adduction moment during gait have been examined in other studies (Fantini Pagani et al., 2010a; Gaasbeek et al., 2007; Pollo et al., 2002; Self et al., 2000; Lindenfeld et al., 1997). There is debate as to the effects on the magnitude during gait; however, studies are difficult to compare as they do not all analyze the same waveform feature to measure magnitude. Some studies measure overall peak, while others differentiate between first and second peak of the knee adduction moment waveform. On top of this, studies make use of different biomechanical models to determine the knee adduction moment or do not report the model at all. It has been shown that discrete peak and mid-stance adduction magnitude values, as well as their timing in the gait cycle, are highly dependent on the biomechanical model used in the analysis (Newell et al., 2008). There also exist associations between osteoarthritis severity and the overall shape of the knee adduction waveform (Astephen et al., 2008) that may influence the timing of peak knee adduction moment. Current research indicates a need to make use of other measures of loading rather than peak values (Maly, 2008; Thorp et al., 2007, 2006). Deluzio and Astephen (2007), Astephen Wilson et al. (2011), Hatfield et al. (2010), and Landry et al. (2007) have shown that Principal Component Analysis (PCA) is an effective tool to describe and compare overall waveform magnitudes between participants.

The discrepancies in results of the studies of the effects of valgus unloader bracing on the knee adduction moment magnitude may also be because the valgus unloader brace does not have the same effect on all patients. In a study by Otis et al. (1996), it was found that participants exhibited varying changes in medial joint load with brace wear. Two of

the six participants in their study exhibited an increase in medial compartment force with the brace on. Although this study was limited due to small sample size, the diversity in the findings suggests that subgroups of participants with varying gait adaptations from wearing a valgus unloader brace may exist and should be further investigated.

Self et al. (2000) and Otis et al. (2000) have argued that other kinetic and kinematic parameters, other than just the abduction moment applied by the brace, may be responsible for its positive treatment of patient pain and function. Some kinematic changes have already been found with brace wear, though the literature is limited and focused primarily on only sagittal plane measures (Ramsey et al., 2007; Richards et al., 2005; Komistek et al., 1999; Davidson et al., 1998). It is therefore necessary to investigate the 3D kinematic and kinetic changes that occur with valgus unloader brace wear to better understand the effects of the brace and, if subgroups of participants do exist, to further clarify if patients exhibit different gait adaptations to brace application.

There are two objectives of this thesis. The first objective is to determine the immediate effect of a custom-fit valgus unloader brace on the overall magnitude of the net external knee adduction moment during gait in individuals with mild to moderate medial compartment knee OA. The second objective is twofold: 1) to identify subgroups whose overall knee adduction moment magnitude i) decreased with brace application, ii) did not change with brace application, and iii) increased with brace application; and 2) to determine differences in 3D knee kinematic and kinetic features after brace application and among subgroups who exhibit varying changes in net knee external adduction moment with the immediate application of a valgus unloader brace.

Chapter 2 is a review of the literature pertinent to this thesis. Each objective has been written as a standalone study and a separate chapter. Chapter 3 addresses the first objective. The second objective, addressed in the methods in Chapter 4, has been prepared as a manuscript to be submitted for publication. Chapter 5 summarizes the conclusions of the study, describes the implications of the findings, and proposes future work to further the understanding of the effects of valgus unloader braces.

CHAPTER 2

BACKGROUND

2.1 Knee Osteoarthritis

2.1.1 *Prevalence and Cost*

The Arthritis Society of Canada (2004) estimates that arthritis currently affects 16% of Canadians. This debilitating disease not only negatively affects patients quality of life, but also has an impact on the Canadian economy. Arthritis has cost Canadians as much as 7.7 billion dollars annually and has cost Americans as much as 1% of their GDP, mostly due to loss of productivity and disability (Public Health Agency of Canada, 2010; Yelin et al., 2004). Nova Scotia has the highest percentage of arthritis cases in Canada, at 23.3%. The prevalence of arthritis also increases with age (The Arthritis Society of Canada, 2004). Between the ages of 30 and 50, the incidence of arthritis increases over five times (Public Health Agency of Canada, 2010). Therefore, as the Canadian population ages, the prevalence of arthritis in Canada is expected to increase to over 20% (Public Health Agency of Canada, 2010).

Osteoarthritis (OA) is the most prevalent type of arthritis in Canada (The Arthritis Society of Canada, 2004). It is characterized by the degradation of articular cartilage and changes in the underlying bone. Osteoarthritis is a leading cause of disability among adults worldwide and can lead to a lower quality of life (Neogi et al., 2011). OA affects three million Canadians (The Arthritis Society of Canada, 2004) and similar numbers are seen in the USA - estimates of the prevalence of knee OA are between 19-28% of adults over 45 years old (Felson et al., 1987; Jordan et al., 2007). As well, the disease affects women

between two and four times more often than men (Buckwalter and Lappin, 2000). The knee is the joint that is most often affected by OA (Buckwalter et al., 2001) and occurs more often in the medial compartment (Felson and Radin, 1994).

2.1.2 Illness and Disease

The Osteoarthritis Research Society International (OARSI) has characterized the difference between OA illness and OA disease (Lane et al., 2011). The disease, structural changes in the joint, and the illness, patient reported symptoms, are not well correlated (Hannan et al., 2000). It is important to understand the difference between OA illness and disease in the study of OA as treatments or compensatory mechanisms may positively alter one, while negatively affecting the other. This is particularly important for therapies that focus on the treatment of pain, as decreased pain can increase walking speeds and joint loading (Schnitzer et al., 1993). This may exacerbate the OA disease process as joint loading has been linked with OA-like deterioration of the articular cartilage in animal models (Roemhildt et al., 2010).

OA illness comprises patient reported symptoms such as the pain, stiffness, and reduced function characterized by OA. These symptoms can decrease patient participation in daily activities and have a negative effect on the overall perceived quality of life of a patient (Hawker et al., 2011, 2010). The primary symptom of OA is pain in the joint. In early stages of the disease, pain may only present after activity, yet in more severe cases, the pain may become constant (Buckwalter et al., 2001). Knee OA patients present to their physician with a poorly localized ache in the knee joint (Buckwalter et al., 2001). In the body of literature about knee OA, the symptoms of OA are generally collected as self-reported measures using various scales (Lane et al., 2011). A commonly used measure is the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) (Bellamy et al., 1988), which measures pain, stiffness, and function in daily activities. However, regardless of scale, all of these measures are still dependent on patient reports. Effects of age or other diseases/stresses in life can have an influence over scores (Lane et al., 2011; Sale et al., 2008).

The disease of OA refers to structural changes within and around the joint. Clinically, these are generally recorded using radiographic images (Buckwalter et al., 2001) and severity is classed using radiographic criteria (Altman et al., 1986) (commonly scored using the Kellgren Lawrence scale (Kellgren and Lawrence, 1957)). Indicators include joint

space narrowing, increased subchondral bone density, and the presence of osteophytes. More severe cases of knee OA present with deformity and malalignment of the joint (Buckwalter et al., 2001).

2.1.3 Biomechanical Changes with Knee OA

Knee OA has an important biochemical and biomechanical etiology (Brandt et al., 2006), although it is generally thought to be mechanically induced (Brandt et al., 2006). It was suggested by Andriacchi et al. (2004) that OA is initiated by a change in knee kinematics, such as from an injury, that shifts the areas of loading in the knee to regions which have not been adapted to high loads. Increased chronic loading on the knees of rabbits has been shown to induce damage in the articular cartilage similar to that seen with OA (Roemhildt et al., 2010). In humans, it is unclear if increased loads will initiate OA, but there it is suggested that a change in loading position may damage areas of cartilage and initiate the disease process (Andriacchi and Mündermann, 2006; Andriacchi et al., 2004). There is evidence, however, that increased loads in the medial compartment are associated with the disease progression of medial compartment OA (Andriacchi and Mündermann, 2006; Mündermann et al., 2005; Miyazaki et al., 2002), characterized by degradation of the cartilage in the medial compartment. The reduction in medial joint space from the cartilage degradation is thought to redistribute the contact loads more heavily toward the medial compartment of the knee (Pollo et al., 2002; Maquet, 1980). The implications of these studies will be discussed more thoroughly in the next section.

2.2 Gait Analysis and Knee OA

Human gait is analyzed to study knee osteoarthritis as it captures the most common daily repetitive function performed by the knee joint. Gait analysis is a method to measure the in vivo function of the knee joint which is related to the structure and health of the knee (Andriacchi et al., 2004). Gait analysis can detect changes that can be associated with the initiation and progression of knee OA (Andriacchi et al., 2004). The study of human gait can provide valuable information on the compensatory mechanisms seen with OA and the effects of treatments (Biden et al., 1990). In modern gait analysis, this information comes in the form of 3D motion of the knee joint, measures of loading on and within the joint, and neuromuscular control patterns of muscles around the knee joint.

In this thesis, gait is evaluated over one gait cycle. The gait cycle represents the foot-strike of one limb (0%) to the next foot-strike of the same limb (100%). Stance phase is identified as the part of the gait cycle in which a participant has their foot in contact with the ground (usually the first 60% of the gait cycle). Stance phase starts with a heel or foot-strike event and ends with a toe-off event. During early stance, weight is shifted from the contralateral limb to the weight bearing limb during a stage called weight acceptance. Swing phase follows stance phase and is the part of the gait cycle during which the limb is lifted off the ground and the contralateral limb is weight bearing. Knee kinematic and kinetic waveforms show variations over the entire gait cycle although some waveforms are of most interest during the load bearing stance phase. When synthesizing information captured using gait analysis, dynamic measures that incorporate the temporal aspect of the mechanical changes should be accounted.

2.2.1 Kinematics

In modern gait analysis, the respective motion of lower limb segments are measured in 3D space. This allows for the examination of gait parameters otherwise not identifiable by the naked eye of a clinician. In this thesis, the relative motion of the thigh and shank segments was evaluated using a joint coordinate system described by Grood and Suntay (1983), that relates the axes of the joint coordinate system to commonly used clinical descriptors of knee joint motion. In this thesis, coordinate systems and motion were defined and recorded for each limb segment using infrared emitting diodes (IRED, Northern Digital Inc. Waterloo, ON) placed on anatomical landmarks (Figure 2.1) and virtual markers on the heel, second metatarsal, medial malleolus, fibular head, tibial tuberosity, lateral and medial epicondyle, and right and left anterior superior iliac spines (Landry et al., 2007). The virtual markers were identified during a quiet standing trial. The locations of IREDs in 3D space were captured at 100 Hz with an OptotrakTM motion capture system (Northern Digital Inc, Waterloo, ON). The subjects wore skin-tight shorts to minimize motion of the greater trochanter marker with respect to the skin.

Using the joint coordinate system developed by Grood and Suntay (1983), the 3D motion at the knee was described about a flexion/extension axis, ab/adduction axis, and internal/external rotation axis (Figure 2.2). The flexion/extension axis was defined by the medial and lateral epicondyle of the femur, internal/external rotation was about the long axis of the shank, from lateral malleolus to the head of the fibula, and the ab/adduction axis was



Figure 2.1: IRED marker set-up. Triads of markers are placed on all limb segments (pelvis, thigh, shank, and foot). Individual markers are placed on the lateral malleolus, greater trochanter, and shoulder (not visible).

a floating axis defined by the cross product of the flexion/extension and internal/external rotation axis vectors.

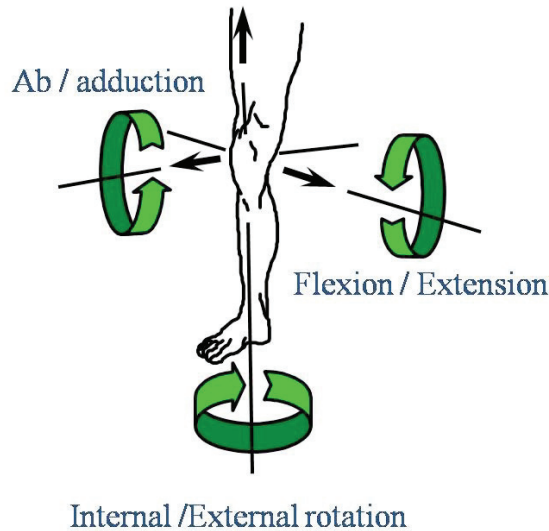


Figure 2.2: Joint coordinate system axes. Flexion/extension axis was defined by the medial and lateral epicondyle of the femur, internal/external rotation was about the long axis of the shank, from lateral malleolus to the head of the fibula, and the ab/adduction axis was a floating axis defined by the cross product of the flexion/extension and internal/external rotation axis vectors (created by Scott Landry, used with permission).

method used to calculate the resultant loads. It is a mathematical method that uses dynamic force and moment balances to calculate net forces and moments acting on the joint in 3D space (Costigan et al., 1992; Vaughan et al., 1992). In this thesis, an inverse dynamics method described by Vaughan et al. (1992) was used. Each limb segment (pelvis, thigh, shank, and foot) was assumed to be separate and rigid. Starting on the most distal segment and working in the proximal direction, a dynamic moment and force balance was performed on each limb segment to determine the reaction forces on proximal segments. The force balance used in inverse dynamics for each limb segment was:

The alignment of the coordinate systems with the directions of motion is important to avoid kinematic crosstalk. Kinematic crosstalk is the projection of off plane motions onto the defined coordinate system and result from misalignment between the coordinate system and in vivo motion (Piazza and Cavanagh, 2000). Kinematic measures calculated with the joint coordinate system have been tested for reliability and at regions of high flexion, rotation and adduction angles have been found to be less reliable (Piazza and Cavanagh, 2000). Therefore, it is important to evaluate the validity of differences detected in these measures during the period of high flexion typically seen in the mid-swing phase of gait (Biden et al., 1990).

2.2.2 Kinetics

Gait analysis also involves measurement of resultant loads on the joints. Inverse dynamics is a commonly employed

$$\sum F = ma = F_{dist} + F_{prox} + mg \quad (2.1)$$

where m is the mass of the limb segment, approximated using anthropometric measures collected in the laboratory and equations described by Vaughan et al. (1992), a is the acceleration of the limb segment, F_{dist} is the force of the distal limb segment acting on the limb segment of interest (the ground reaction force in the case of the foot), F_{prox} is the force of the proximal limb segment acting on the limb segment of interest, and g is the acceleration due to gravity.

The moment balance was calculated about the centre of mass of the limb. The moment balance used in inverse dynamics for each limb segment was:

$$\sum M = I\alpha = M_{dist} + M_{prox} + F_{dist} * r_{dist} + F_{prox} * r_{prox} \quad (2.2)$$

where I is the moment of inertia of the limb segment, approximated using anthropometric measures collected in laboratory and equations described by Vaughan et al. (1992), α is the angular acceleration of the limb segment, M_{dist} is the moment of the distal limb segment acting on the limb segment of interest (ground reaction moment in the case of the foot), M_{prox} is the moment of the proximal limb segment acting on the limb segment of interest, r_{dist} is the moment arm between the line of action of F_{dist} and the limb segment center of mass, and r_{prox} is the moment arm between the line of action of F_{prox} and the limb segment center of mass. A 2D sagittal plane representation of the inverse dynamics method on the knee is shown in Figure 2.3.

In this thesis, a ground embedded AMTITM force platform (Advanced Mechanical Technology Inc, Watertown, MA) was used to calculate the ground reaction forces and moments acting on the foot of the limb of interest during the stance phase of gait. When the moment and force balances were performed on the shank segment over the entire gait cycle, F_{prox} and M_{prox} represented the net external force and moment at the knee joint. These were then projected onto the axes defined by the joint coordinate system to describe knee loading. All moments were normalized to body mass (Landry et al., 2007) and waveforms were normalized to 101 data points from foot-strike to foot-strike representing 100% of the gait cycle.

Although inverse dynamics can provide information on loading of the joints, it must

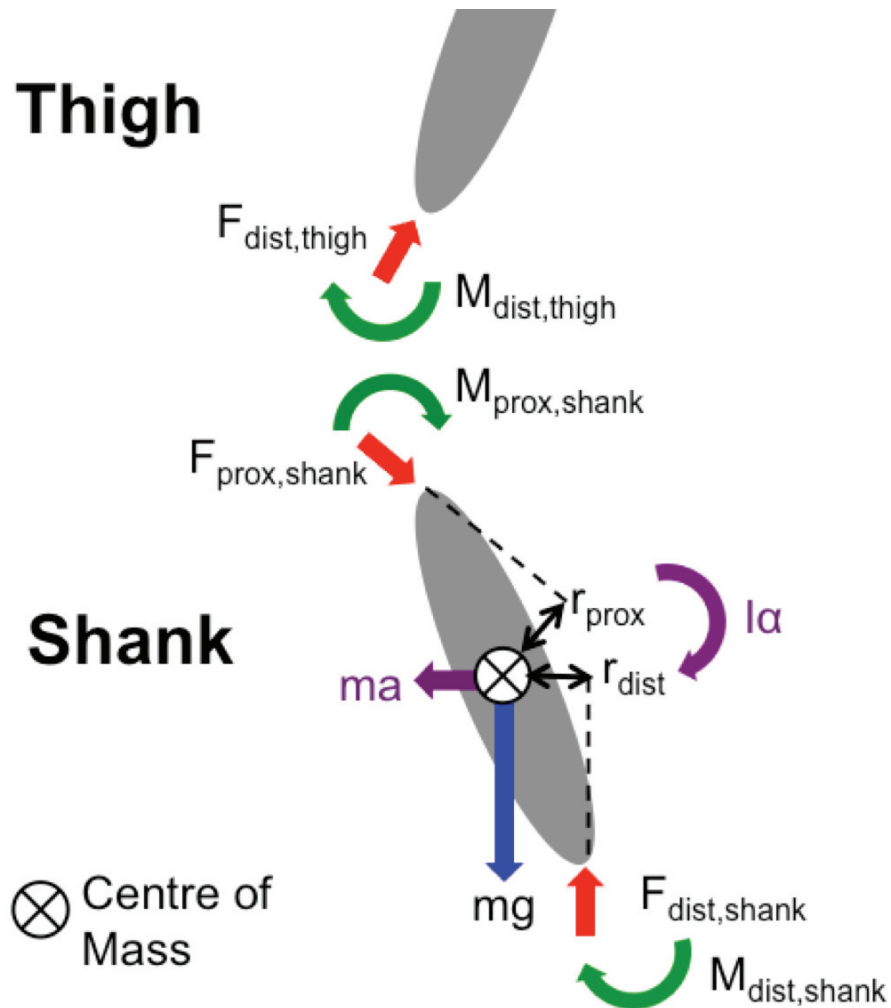


Figure 2.3: 2D inverse dynamics. A 2D sagittal plane representation of the forces and moments that are involved in the inverse dynamics process at the knee, where m is the mass of the shank, a is the acceleration of the shank, g is the acceleration due to gravity, $F_{dist, shank}$ is the force of the foot acting on the shank at the ankle, $F_{prox, shank}$ is the force of the thigh acting on the shank at the knee, $F_{dist, thigh}$ is the force of the shank acting on the thigh at the knee, I is the moment of inertia of the shank, α is the angular acceleration of the shank, $M_{dist, shank}$ is the moment of the foot acting on the shank at the ankle, $M_{prox, shank}$ is the moment of the thigh acting on the shank at the knee, $M_{dist, thigh}$ is the moment of the shank acting on the thigh at the knee, r_{dist} is the moment arm between the line of action of $F_{dist, shank}$ and the shank center of mass, and r_{prox} is the moment arm between the line of action of $F_{prox, shank}$ and the shank center of mass.

be recognized that resultant moments and forces are a *net* effect of all the external forces and moments on the joint. Inverse dynamics does not describe the distribution of load within the joint nor attribute values of moment and force to individual passive and active structures. This is of importance in the case of knee osteoarthritis as the agonist-antagonist contraction of muscles creates forces that may work in opposing directions (Hubley-Kozey et al., 2008). These forces would add to the loading of the knee joint, but would be nullified in an inverse dynamics analysis.

Forward dynamics is another method used to describe knee loading. It is generally used to calculate motion from internal forces and moments (Otten, 2003). In biomechanics, the internal forces are known either through estimation and assumption or through a direct measure, such as implanting a force transducer in a tendon. Directly measuring individual soft tissue or contact forces is regularly done in animal studies using implanted devices (Takahashi et al., 2010; Rijkelijkhuisen et al., 2007); however, this is generally not feasible for experimentation on humans. More commonly, loads in individual structures in human joints are estimated through modelling (Buchanan et al., 2004; De Vita and Hortobagyi, 2001) or measured with instrumented prosthetic implants already prescribed to the patient (Erhart et al., 2010; Zhao et al., 2007).

2.2.3 *Neuromuscular Activation*

Beyond understanding the forces imposed kinetically, the study of forces in gait analysis can include the observation and analysis of muscle activation. Muscles have been shown to contribute 3-5 times the force of body weight in overall knee loading (Taylor et al., 2004). Electromyography (EMG) is a method by which the action potential in muscles is recorded using either surface or intramuscular fine wire electrodes (Cresswell et al., 1994; Hubley-Kozey et al., 2008). Muscle activation can be included in forward dynamics analyses to estimate muscle forces (Buchanan et al., 2004), or they can be examined on their own (Hubley-Kozey et al., 2008). Not only does EMG provide more information about the loading of the joint, it also provides information about compensatory mechanisms seen with OA. Hubley-Kozey et al. (2008), Childs et al. (2004), and Benedetti et al. (1999) have all shown that there is increased co-activation of the musculature surrounding the knee during gait in patients with knee OA. It is suggested that this compensatory mechanism increases stiffness in the joint in response to pain and instability.

2.2.4 *Alterations in Medial Compartment Loading with Knee OA During Gait*

As presented earlier, there are biomechanical changes associated with knee OA. Many of these have been made apparent in vivo using gait analysis. In particular, knee loading has been associated with the progression and severity of knee OA. This section will discuss this link.

As OA most often occurs in the medial compartment of the knee, studying the associations between OA and medial joint load has been of particular importance to knee OA researchers. However, directly measuring this load is generally difficult as it would involve implanting instrumentation into the knee joint. Erhart et al. (2010) and Zhao et al. (2007) have correlated medial joint contact forces in an instrumented prosthesis with the peak net external knee adduction moment measured using the inverse dynamics method. Therefore, the knee adduction moment is often used as a surrogate measure of medial joint load in gait literature (Gaasbeek et al., 2007; Mündermann et al., 2005).

High knee adduction moments have been associated with the progression and severity of knee OA. Patients with high medial compartment knee OA radiographic severity grades have higher peak knee adduction moments than patients with lesser radiographic severity (Mündermann et al., 2005). As well, a high knee adduction moment in patients with less severe OA may be indicative of future progression. In a study by Miyazaki et al. (2002), patients with high knee adduction moments were more likely to exhibit radiographic OA progression over six years. As the knee adduction moment is associated with the forces in the medial compartment of the knee joint, increased knee adduction moments have become an indicator of patients' risk of progression of medial compartment knee OA (Andriacchi and Mündermann, 2006).

2.2.5 *Alterations in 3D Kinetics and Kinematics with Knee OA During Gait*

Besides changes in medial joint loading, OA has been shown to alter 3D kinetic and kinematic measures of the knee during gait. Studies have demonstrated the existence of differences in 3D loading among asymptomatic and OA participants, as well as between varying levels of severity (Astefhen et al., 2008; Landry et al., 2007; Gok et al., 2002; Baliunas et al., 2002; Hurwitz et al., 2002; Kaufman et al., 2001; Sharma et al., 1998).

However, very few of these studies were longitudinal in design and conclusions about progression were often drawn using cross sectional studies of OA populations with different severity levels. The limits of these comparisons must be considered when evaluating possible biomechanical factors of progression. In the sagittal plane, a reduced knee flexion moment magnitude in early stance (Astefhen et al., 2008; Landry et al., 2007; Kaufman et al., 2001) and a reduced extension moment in late stance (Gok et al., 2002; Baliunas et al., 2002) have been shown in an OA population. In one study, participants with severe OA experienced a knee flexion moment throughout the entire stance phase, without an extension moment at mid-late stance (Astefhen et al., 2008). There is disagreement in the literature as to the effects of OA on the knee rotation moment (Landry et al., 2007; Gok et al., 2002; Kaufman et al., 2001); however, this may be due to differences between the waveform analysis techniques used in these studies as changes in timing and magnitude of the knee rotation moment have been shown with increasing OA severity (Astefhen et al., 2008).

Knee OA has also been associated with changes in 3D motion of the knee joint as well as loading (Kuroyanagi et al., 2011; Astefhen et al., 2008; Landry et al., 2007; Chang et al., 2004; Gok et al., 2002; Kaufman et al., 2001). Moderate OA participants exhibit less flexion throughout the gait cycle than controls (Landry et al., 2007; Gok et al., 2002; Kaufman et al., 2001) and severe OA participants have been shown to have less extension in late stance phase (Astefhen et al., 2008). In the frontal plane, the existence of varus thrust, an abrupt lateral motion of the knee in early stance during weight acceptance (Chang et al., 2004, 2010), has been associated with the progression of knee OA in a longitudinal study (Chang et al., 2004) and its magnitude has been linked with the magnitude of the knee adduction moment during gait, an indicator of disease progression risk (Kuroyanagi et al., 2011). No studies have evaluated the effect of the knee rotation angle on knee OA progression.

A recent study by (Henriksen et al., 2011) showed that the knee rotation moment peak is associated with the knee adduction moment peak and suggests that the role of knee rotation moment in the disease process of knee OA should be investigated more thoroughly. As more studies evaluate kinematic and kinetic changes with knee progression using a longitudinal design, other biomechanical indicators of the risk of knee OA progression may become apparent.

2.2.6 *Principal Component Analysis*

Although modern gait analysis can provide an extensive amount of information on knee joint motion, loading, and muscle activation, comparing gait information between subjects can be difficult. Commonly, gait researchers extract discrete variables, such as range of motion or peaks, as is often the case with the knee adduction moment (Gaasbeek et al., 2007; Mündermann et al., 2005). However, some gait variables show changes in shape over the entire gait cycle as well as changes in magnitude with increasing OA severity (Astefhen et al., 2008). It has been shown that discrete peak and mid-stance adduction magnitude values as well as their timing within the gait cycle (foot-strike or toe-off) are highly dependent on the chosen biomechanical model (Newell et al., 2008). Current research has indicated a need to explore other measures of loading than peak values (Maly, 2008; Thorp et al., 2007, 2006).

Principal Component Analysis (PCA) is a pattern recognition technique that can reduce gait waveform data to be easily interpretable and comparable between subjects. Deluzio and Astefhen (2007), Astefhen Wilson et al. (2011), Hatfield et al. (2010), and Landry et al. (2007) have shown that PCA is an effective tool to describe and compare gait waveforms between participants. For each waveform type, a data matrix, \mathbf{X} , is created that includes the gait cycle data of a sample group. The rows represent observations of the respective gait measure for participants and columns represent the time points of the time normalized gait cycle. Eigenvectors, referred to as principal components (PCs), are extracted from the covariance matrix of \mathbf{X} , and these describe the primary patterns of variability in the waveform data set (Figure 2.4). A PC-score is calculated for each participant for the waveform of interest. The PC-score is the projection of the participant's data onto the PC, calculated as $\text{PC-score} = (\mathbf{X})\mathbf{U}$, where \mathbf{U} is the matrix of the reduced number of eigenvectors (Landry et al., 2007). An individual's PC-score represents how closely an individual's waveform pattern corresponds to the pattern described by that PC. To better interpret the PCs and the meaning of the scores, the relative importance of the PC at each portion of the gait cycle is often evaluated. The percent variation explained, *Percent Variation Explained(i,j)*, is a measure of the amount of variation in the participant waveforms at each time point that is explained by the PC, PC_j , at that time point, i (Deluzio and Astefhen, 2007).

PCA provides information about the dynamic measures associated with gait analysis.

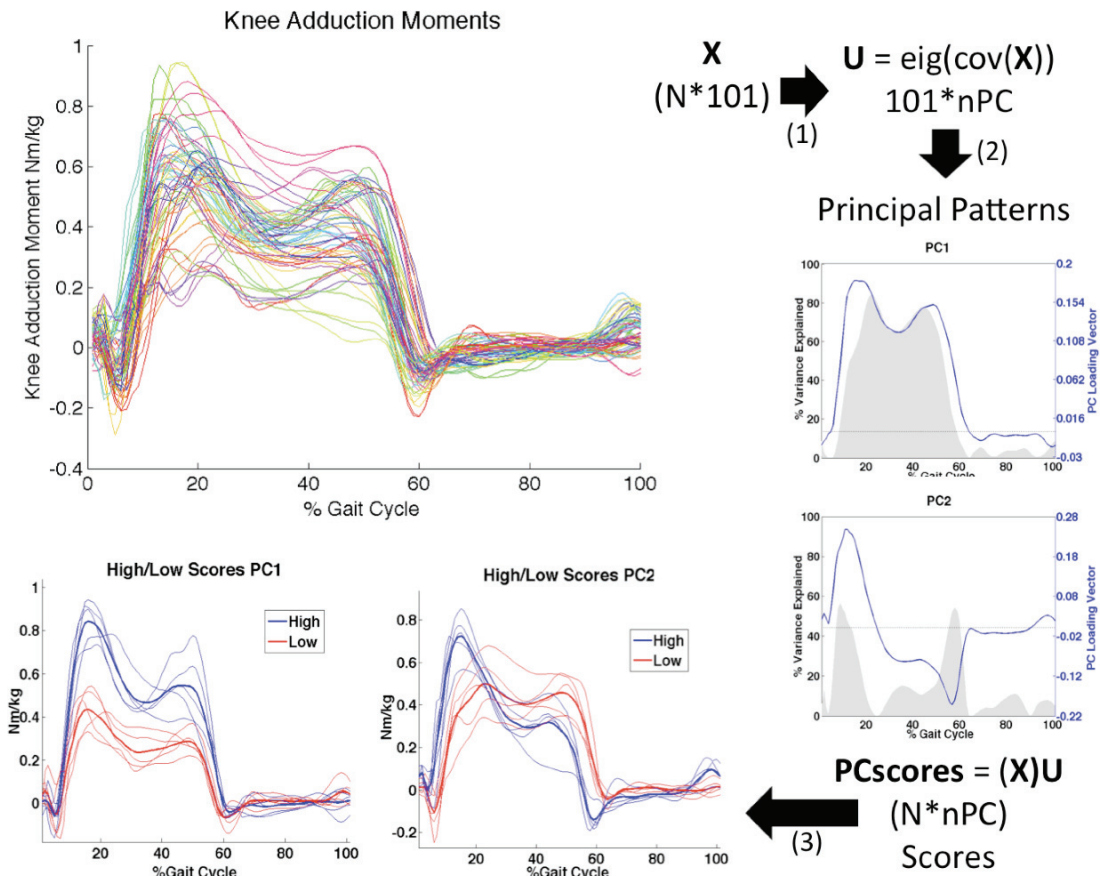


Figure 2.4: An example of Principal Component Analysis (PCA) on the knee adduction moment. \mathbf{X} is a matrix of all participant knee adduction moment waveforms (101 time points \times N participants) (1). The PCs are the eigenvectors of the covariance matrix of \mathbf{X} (2). $n\text{PC}$ principal components are extracted from the covariance matrix ($n\text{PC} = 2$ in this example). The percent variance explained by each PC is represented by the shaded area under each PC curve (2). Each knee adduction waveform is given a score calculated by $\text{PC-score} = (\mathbf{X})\mathbf{U}$, where \mathbf{U} is the matrix of the two eigenvectors (3). Examples of five waveforms with high and low PC score (5th and 95th percentile) and the average of the waveforms for each PC are shown (high = blue, red = low (3)).

It describes both magnitude and temporal features of gait waveforms. It has been used to evaluate muscle activation patterns, as it has the ability to quantify amplitude and shape characteristics of EMG (Hubley-Kozey et al., 2008). PCA has been able to discern differences between OA populations differing in severity (Asthephen et al., 2008). PCA has also been shown to have limited sensitivity to different biomechanical models used to define osteoarthritic gait patterns (Newell et al., 2008). As well, the first principal component of the knee adduction moment, which typically describes the overall magnitude of the knee adduction moment during the stance phase (Hatfield et al., 2010; Asthephen Wilson et al., 2011; Landry et al., 2007), has been correlated with the area under the curve measurement (Newell et al., 2008). This is similar to the knee adduction angular impulse measured by Thorp et al. (2007, 2006) which is another representation of joint loading over the entire stance phase. This thesis will use PCA to describe and compare gait waveforms between participants.

2.3 Treatments

Treatments for knee osteoarthritis vary depending on the severity of the disease. The Osteoarthritis Research Society International (OARSI) recommends that patients who cannot achieve pain relief and functional improvement from more conservative treatments, should be considered for a joint replacement (Zhang et al., 2008). In 2008-2009, over 47 000 knee replacements were performed in Canada (Canadian Institute for Health Information, 2010). The total number of hip and knee replacements in this timeframe was slightly under 78 000, less than three percent of the estimated three million Canadians living with osteoarthritis. Access to knee replacement surgery is therefore a barrier. In 2006-2007, the median wait time for a knee replacement surgery was 169 days (Canadian Institute for Health Information, 2009). Younger and younger patients are getting total knee replacements (Canadian Institute for Health Information, 2009; Jain et al., 2005). Over the last two decades, patients aged 40-55 have undergone a dramatically greater proportion of the knee replacement surgeries performed in Canada and the US. Knee replacement joints have a limited lifetime and knee replacement revision rates are expected to increase (Kurtz et al., 2007). It is projected that the capacity to provide orthopaedic treatments will only decline in the future (Dunbar et al., 2009).

Because only a small number of patients will receive surgical intervention, there is

an increasing demand for non-operative, conservative treatments of OA (Buckwalter et al., 2001). Common pharmacological treatments are orally ingested anti-inflammatory medication and corticosteroid or hyaluronan injections. Orally ingested tablets to control knee pain are the most common treatment used in the management of knee OA; however, patients show a strong desire for research into alternative treatments (Tallon et al., 2000). There is also evidence that nonsteroidal anti-inflammatory drugs decrease pain yet increase walking speed and loading at the joint (Schnitzer et al., 1993), which may exacerbate OA progression. As younger and younger patients are receiving surgical intervention, there is a further need for treatments that will slow the progression of knee OA. Orthotic aids such as heel wedges, canes, and braces have the common aim of reducing the forces in the knee joint that occur with daily activity, such as walking (Buckwalter et al., 2001). The focus of this study is on bracing as it aims to not only reduce pain but to improve the mechanical environment of the knee as well.

2.4 Bracing

OARSI recommends valgus unloader braces to reduce pain, improve stability, and reduce risk of falling for knee OA patients with mild to moderate varus or valgus instability (Zhang et al., 2008). Although several brands of valgus unloader braces exist, they are generally all designed to apply an abduction moment to the knee through adjustments to a hinge connecting the thigh and shank segments (Pollo et al., 2002; Self et al., 2000) (Figure 2.5). Unloader bracing is also intended to reduce medial compartment pain by reducing medial compartment force through the applied external abduction moment at the knee (Pollo et al., 2002). As medial compartment forces have been linked with OA progression, a reduction of medial compartment load may slow the progress of the disease.

2.4.1 Bracing as a Treatment of Illness

Valgus unloader bracing is effective in reducing pain in patients with medial compartment knee osteoarthritis and adducted alignment (Gaasbeek et al., 2007; Richards et al., 2005; Kirkley et al., 1999; Komistek et al., 1999; Lindenfeld et al., 1997). The effect of pain relief was seen immediately (Komistek et al., 1999), at six weeks (Gaasbeek et al., 2007; Lindenfeld et al., 1997), and after six months of wear (Richards et al., 2005; Kirkley et al., 1999). Although pain relief was seen across all studies, Komistek et al. (1999) noted

that the three of 15 participants in his study who did not have pain relief were also obese and blamed poor brace fixation as the cause. As well as pain, improved physical function is also often reported by knee OA patients as a result of brace wear (Gaasbeek et al., 2007; Richards et al., 2005; Kirkley et al., 1999; Komistek et al., 1999; Lindenfled et al., 1997). It is clear that valgus unloader bracing results in a better quality of life, but no studies occurred over a long enough time period to study the relationship between self-reported outcomes and disease progression. It is important to remember that although bracing successfully treats the illness of OA, this is not proof that it treats the disease.

2.4.2 *Bracing as a Treatment of Disease*



Figure 2.5: Valgus unloader brace (Breg brand). The thigh and shank segments can be locked into an abducted position.

The immediate effects of valgus unloader bracing on the net external knee adduction moment have been examined in several studies (Fantini Pagani et al., 2010a; Gaasbeek et al., 2007; Pollo et al., 2002; Self et al., 2000; Lindenfled et al., 1997) after two - six weeks of wear; however, there are discrepancies between studies as to its effect on the magnitude of the knee adduction moment during gait. The discrepancies between studies may exist because there are differences in which measures were used to represent knee adduction moment magnitude. The peak stance phase of the net external knee adduction moment is the most commonly used measure (Gaasbeek et al., 2007; Pollo et al., 2002; Lindenfled et al., 1997); however, other measures of knee adduction moment magnitude are used as well. Self et al. (2000) measured several points during 15-30% of stance phase and found a reduction at 20 and 25% only. In two recent studies, Fantini Pagani et al. (2010a) and Toriyama et al. (2011) evaluated both first and second peaks of the external knee adduction moment during stance. An understanding of the loading at foot-

strike and toe-off provides a better understanding of the effect of the brace throughout the gait cycle; however, the information that can be drawn from this analysis is still limited as changes in the overall shape of the knee adduction moment waveform have been associated

with differences in knee OA severity (Aststephen et al., 2008).

Even between the studies which evaluated the peaks in the knee external knee adduction moment, there is still debate as to the effects of brace application on the external knee adduction moment. Gaasbeek et al. (2007) and Pollo et al. (2002) did not find a reduction in peak external knee adduction moment with brace application, while Lindenfeld et al. (1997) did find a significant reduction in the peak external knee adduction moment. Fantini Pagani et al. (2010a) found a reduction in the second peak, but not the first, while Toriyama et al. (2011) found a reduction in the first peak, but not the second. These discrepancies may exist because of differences in the choice of biomechanical model used to calculate the knee adduction moment. The choice of biomechanical model used in a study can have an effect on the relationship between the first and second peak and mid-stance adduction values (Newell et al., 2008). Of the five studies which evaluated the peaks in the external knee adduction moment waveform, the biomechanical model was only reported in three and those studies did not use the same model.

Whether or not the brace affected walking speed may also be the cause of some of the discrepancies in the literature regarding the knee adduction moment magnitude. Increases in walking speed have been shown to affect the knee adduction moment magnitude (Landry et al., 2007). Gaasbeek et al. (2007), Pollo et al. (2002), and Lindenfeld et al. (1997) did not show a reduction in walking speed between conditions, however Fantini Pagani et al. (2010a) and Toriyama et al. (2011) did find a difference with brace application. Self et al. (2000) did not measure gait speed and suggests this as a possible reason for increased knee adduction moments.

To clarify the effects of a valgus unloader brace on the knee adduction moment, there is a need to evaluate the change in knee adduction moment magnitude incurred with brace application over the entire stance phase. Fantini Pagani et al. (2010a) did evaluate a measure of the overall change in knee adduction moment magnitude over the entire stance phase; however, the brace used in that study was much larger than what is typically available on the market as the thigh and shank segments were extended over the entire length of the leg. That study measured the knee adduction angular impulse, a measure of the area under the knee adduction moment curve that represents loading over the entire stance phase (Thorp et al., 2006), and was able to discern a reduction in impulse caused by the brace. No studies have used PCA to measure the change in overall magnitude of the

external knee adduction moment over the entire gait cycle with valgus unloader brace wear. Using PCA would provide an overall measure of the magnitude of the knee adduction moment over the entire stance phase. As well, PCA is an appropriate tool for the evaluation of the knee adduction moment as it has limited sensitivity to different biomechanical models used to define the gait patterns seen in participants with knee osteoarthritis (Newell et al., 2008). In particular, PC1 had the least sensitivity to the chosen biomechanical model in an evaluation of the knee adduction moment (Newell et al., 2008) and it is this principal component which generally represents the knee adduction moment magnitude (Asthephen Wilson et al., 2011; Hatfield et al., 2010; Landry et al., 2007). This thesis will evaluate the effect of commonly prescribed valgus unloader braces on the knee adduction moment magnitude over the entire stance phase of gait.

Patients may not all react the same way to brace wear. Using an analytical model to determine medial compartment force, Otis et al. (1996) showed that not all patients respond in the same way to valgus unloader brace wear. Some participants even exhibited an increase in medial compartment forces during the stance phase of gait. Different gait adaptations with varying effects on medial joint loading between patients may be another reason for the uncertainty in the literature regarding the change in knee adduction moment magnitude with brace wear. (Otis et al., 1996) was limited in sample size and the existence of groups of participants exhibiting different gait populations within a larger sample should be investigated.

Several studies have evaluated the abduction moment applied to the osteoarthritic knee by the brace using an instrumented brace (Fantini Pagani et al., 2010a; Pollo et al., 2002; Otis et al., 2000, 1996). The size of the abduction moment applied by the brace is small (Pollo et al., 2002; Otis et al., 2000, 1996) and less than 10% of the knee adduction moment magnitudes typical in the mild to moderate OA population (Asthephen et al., 2008; Landry et al., 2007) . Studies which measure the brace abduction moment usually include it in their analysis of medial joint load (Fantini Pagani et al., 2010a; Pollo et al., 2002; Otis et al., 1996). Fantini Pagani et al. (2010a) subtracted the brace abduction moment waveform from the external knee adduction moment to represent medial joint load. Pollo et al. (2002) used an analytical model to estimate the medial compartment contact force. Although this model included changes in the flexion/extension moment that occurred with brace wear, the model in the frontal plane only included the external

knee adduction moment, the brace abduction moment, and the contact loads in the knee. The addition of the brace is comparable to the addition of another passive structure in the knee, such as a ligament, and therefore the moment created by the brace is comparable to the moment created by a ligament; however, these studies did not account for the other changes in the passive and active structures surrounding the knee (Figure 2.6). It has been shown that the net force on a joint is largely caused by forces developed in the musculature, instead of external body weight forces (Taylor et al., 2004). The studies that used an instrument brace assumed that there were no changes to the distribution of joint load within the knee internal structures other than the abduction moment incurred by the brace. This assumption must be examined carefully, particularly as unloader bracing reduces vastii and hamstring co-contraction associated with knee OA during weight acceptance in gait after as little as two weeks of wear (Ramsey et al., 2007).

The net external knee adduction moment is the sum of all external moments acting on the knee in the frontal plane and is therefore equal to the sum of all the internal moments created by contact loads and passive and active structures within the knee. Because it is related to medial compartment force and represents the net effect of all external forces and moments at the knee, this thesis will analyze the effect on the net external knee adduction moment.

In the literature, there is still debate regarding whether or not the brace reduces the

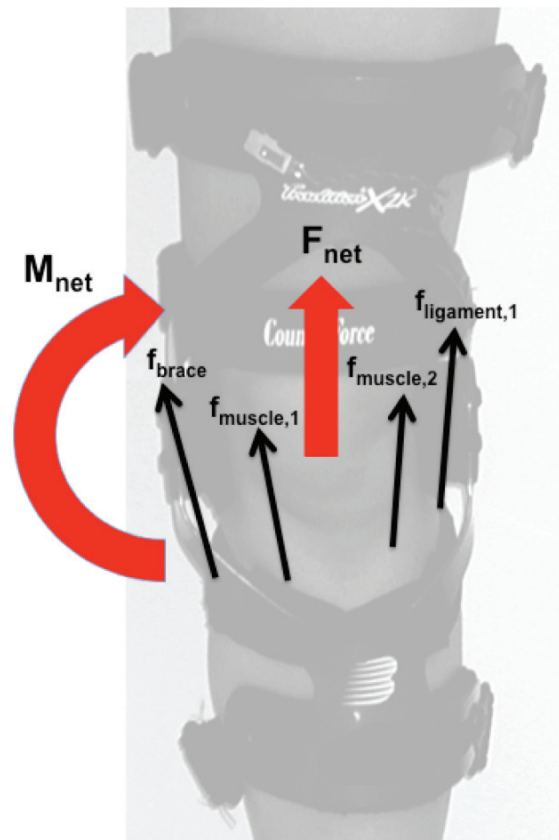


Figure 2.6: The forces and moments acting on the shank from the passive and active structures around the knee joint sum to the net external force and moment (F^{net} , M^{net}). The brace is only one of many structures over which the overall overall load is distributed, and in the case of muscular co-contraction, sometimes in opposing directions.

medial joint load of all patients and successfully treats disease progression. However, as discussed earlier, self-reported pain and function have been repeatedly shown to improve; therefore, Self et al. (2000) and Otis et al. (2000) argue that other kinetic and kinematic parameters more than just the abduction moment from the brace may be responsible for its success in treating OA illness. The effects of the brace on the 3D kinematics and kinetics of the knee joint have been investigated in the literature.

In the sagittal plane, the brace may inhibit the range of flexion and extension. It has been shown to reduce flexion during swing phase (Richards et al., 2005), but inhibit extension at mid-stance (Davidson et al., 1998), and before foot-strike (Gaasbeek et al., 2007). Reduced flexion may hinder foot clearance and create a shorter stride length (Richards et al., 2005). The brace has also been shown to increase the flexion moment at early stance and reduce the extension moment at late stance (Toriyama et al., 2011). A slight increase in walking speed (4 cm/s) was suggested as the cause of the changes in flexion/extension moment seen in that study; however, this increase in gait speed was much smaller than that seen by Landry et al. (2007) in moderate OA participants between self-selected and fast walking (43 cm/s) which caused similarly sized changes in knee flexion moment magnitudes.

Off sagittal plane angles have not been thoroughly reviewed in the literature, other than two studies (Ramsey et al., 2007; Davidson et al., 1998). In the transverse plane, Davidson et al. (1998) found that the thigh was more externally rotated, and the shank was more internally rotated at foot-strike and toe-off. However, this analysis was performed in the global frame and on each limb segment separately so it is not possible to know if participants showed overall internal rotation of the shank with respect to the thigh or if these were two separate gait patterns exhibited by different subjects. Little is known within the research community as to the relationship between rotation angle and knee OA; however, it has been suggested that a change in the rotational characteristics at the knee may shift loading to previously unloaded areas of cartilage and initiate knee OA (Andriacchi and Mündermann, 2006).

In the frontal plane, condylar separation angle of the knee and joint space between the tibial plateau and medial and lateral condyles is increased at foot-strike when wearing an unloader brace (Komistek et al., 1999). Davidson et al. (1998) showed that the shank was more abducted with brace wear at toe-off and Ramsey et al. (2007) showed reduced frontal plane excursions toward knee adduction from foot strike to the time of peak flexion.

Although these results seem to agree and may be a positive change in terms of progression (Chang et al., 2004), Ramsey et al. (2007) found no change in the peak knee adduction angle which shows that the brace did not alter frontal plane alignment throughout the entire gait cycle. Participants with reduced excursions had the most medial joint space narrowing and were classified with "greater stiffening of the knee" as they had reduced flexion during weight acceptance.

None of these studies have examined all 3D kinematics and kinetics over the entire gait cycle. Nor have any studies tried to relate changes in 3D kinematics and kinetics to a change in knee adduction moment with brace wear. Determining if this relationship exists would be an important step toward clarifying:

- the mechanism by which the brace affects the knee adduction moment,
- if there are other mechanisms by which the brace might be causing positive self-reported outcomes,
- if there are groups of patients who exhibit demographic, kinematic, and/or kinetic traits that are more likely to receive the beneficial biomechanical effects of brace wear that are related to disease progression.

The majority of the bracing literature has examined the short term effect of brace application within two - six weeks of brace wear (Fantini Pagani et al., 2010a; Gaasbeek et al., 2007; Ramsey et al., 2007; Pollo et al., 2002; Self et al., 2000; Lindenfeld et al., 1997) and one study examined the effect of the brace after first application (Komistek et al., 1999). Studies have also examined long-term effects of brace wear on gait mechanics after six months of wear (Richards et al., 2005; Hewett et al., 1998). In studying the gait mechanics, it is important to evaluate both immediate and long term effects of brace application, as well as the original gait mechanics of the patients. Each time point can provide information that can help in understanding the potential mechanisms of change and effects of bracing. Gait analysis after long term use can show whether the brace can more permanently alter gait and whether these changes are associated with benefits in terms of pain and disease progression. Gait analysis after immediate application will show the mechanisms through which the brace imposes changes in knee function on patients and may clarify how the brace effects any long term changes in knee function. An understanding of the original gait mechanics of patients could help predict who will most benefit from brace wear if

differences between patient outcomes exist. This thesis evaluated the immediate effects of brace application; however, this thesis is part of a larger study which evaluates gait biomechanics at all three time points.

In summary, there remains a need to investigate the effect of valgus unloader bracing on the knee adduction moment magnitude over the entire stance phase of the gait cycle. Other 3D kinetic and kinematic measures of knee joint function should be evaluated to better understand the effects of the brace. There is evidence, although limited in sample size, that the brace may have varying effects on the medial compartment force in the knee. This should be further investigated using a larger sample size. As well, other 3D kinematic and kinetic differences between participant groups should be examined in order to further clarify if patients exhibit different gait adaptations in response to valgus unloader brace application.

CHAPTER 3

DOES THE VALGUS UNLOADER BRACE AFFECT THE EXTERNAL KNEE ADDUCTION MOMENT MAGNITUDE?

3.1 Introduction

Osteoarthritis is a leading cause of disability among adults worldwide (Neogi et al., 2011). Osteoarthritis affects approximately one tenth of all Canadians (Health Canada, 2003) and this number, along with associated health effects (The Arthritis Society of Canada, 2004), is expected to rise as the population of Canada ages (Public Health Agency of Canada, 2010). Similar numbers are seen in the USA - estimates of the prevalence of knee OA are between 19-28% of adults over 45 years old (Felson et al., 1987; Jordan et al., 2007).

In the knee, the medial compartment is more commonly affected than the lateral (Felson and Radin, 1994). In the medial compartment, the cartilage degrades and there is a decrease in the medial joint space (Buckwalter et al., 2001). This redistributes the contact loads in the knee toward the medial compartment (Maquet, 1980). Knee joint loading has been identified as a factor in disease progression (Aststephen et al., 2008; Andriacchi and Mündermann, 2006). The external knee adduction moment magnitude has been associated with medial compartment contact force in the knee and is therefore often considered a surrogate measure of medial compartment load (Erhart et al., 2010; Zhao et al., 2007). Both severity and progression of medial compartment knee osteoarthritis have been linked

to high peak net external knee adduction moments (Andriacchi and Mündermann, 2006; Mündermann et al., 2005; Miyazaki et al., 2002).

Valgus unloader braces are prescribed to mild to moderate medial compartment knee osteoarthritis patients to relieve pain and increase stability (Zhang et al., 2008). They have been shown to improve self-reported pain and function (Gaasbeek et al., 2007; Richards et al., 2005; Kirkley et al., 1999) and patients walked faster with and without the brace after extended periods of brace wear (Gaasbeek et al., 2007). Valgus unloader braces are purported to apply an abduction moment at the knee to reduce the medial compartment load associated with OA (Pollo et al., 2002). As the brace is locked in more abduction than that of the patient, the brace was found in one study to induce condylar separation at foot-strike (Komistek et al., 1999) and this may reduce the medial joint load.

The immediate effects of valgus unloader bracing on the net external knee adduction moment have been examined in several studies (Fantini Pagani et al., 2010a; Gaasbeek et al., 2007; Pollo et al., 2002; Self et al., 2000; Lindenfled et al., 1997) after two - six weeks of wear; however, there are discrepancies between studies as to its effect on the magnitude of the knee adduction moment during gait. The discrepancies between studies may exist because there are differences in which measures were used to represent knee adduction moment magnitude. The peak stance phase of the net external knee adduction moment is the most commonly used measure (Gaasbeek et al., 2007; Pollo et al., 2002; Lindenfled et al., 1997); however, other measures of knee adduction moment magnitude are used as well. Self et al. (2000) measured several points during 15-30% of the stance phase and found a reduction at 20 and 25% only. In two recent studies, Fantini Pagani et al. (2010a) and Toriyama et al. (2011) evaluated both foot-strike and toe-off peaks of the external knee adduction moment during stance. Evaluating the adduction moment at both peaks provides a better understanding of the effect of the brace throughout the gait cycle than just evaluating one peak; however, the information that can be drawn from this analysis is still limited as changes in the overall shape of the knee adduction moment waveform have been associated with differences in knee OA severity (Aststephen et al., 2008).

Even between the studies which evaluated the peaks in the knee external knee adduction moment, there is still debate as to the effects of brace application on the external knee adduction moment. Gaasbeek et al. (2007) and Pollo et al. (2002) did not find a reduction in

peak external knee adduction moment with brace application, while Lindenfeld et al. (1997) did find a significant reduction in the peak external knee adduction moment. Fantini Pagani et al. (2010a) found a reduction in the second peak, but not the first, while Toriyama et al. (2011) found a reduction in the first peak, but not the second. These discrepancies may exist because of differences in the choice of biomechanical model used to calculate the knee adduction moment. The choice of biomechanical model used in a study can have an effect on the relationship between the first and second peak and mid-stance adduction values (Newell et al., 2008). Of the five studies which evaluated the peaks in the external knee adduction moment waveform, the biomechanical model was only reported in three and those studies did not use the same model. As well, subject factors such as walking speed and OA severity are not consistent throughout the literature and may be another cause of some of the discrepancies.

Current research indicates a need to explore other measures of the knee adduction moment beside peak values to capture a value of loading over the entire gait cycle (Maly, 2008; Thorp et al., 2007, 2006). Furthermore, there are associations between osteoarthritis severity and the overall shape of the knee adduction waveform that may be lost using peak values (Astefhen et al., 2008). One study has evaluated a measure of the overall change in knee adduction moment magnitude over the entire stance phase (Fantini Pagani et al., 2010a). This study measured the knee adduction angular impulse, a measure of the area under the knee adduction moment curve that represents loading over the entire stance phase (Thorp et al., 2006), and was able to discern differences caused by the brace; however, the brace used in that study was much larger than what is typically available. Therefore, there remains a need to evaluate the overall magnitude of the knee adduction waveform over the entire stance phase after application of a more commonly prescribed valgus unloader brace. Deluzio and Astefhen (2007), Astefhen Wilson et al. (2011), Hatfield et al. (2010), and Landry et al. (2007) have shown that Principal Component Analysis (PCA) is an effective tool to describe and compare overall waveform magnitudes as well as temporal features in waveforms between participants. Only the magnitude of the knee adduction moment is of interest in this chapter; however, pattern changes in the knee adduction moment waveform will be investigated in the next chapter using PCA. Therefore to be consistent with analysis techniques between chapters, PCA will be used to evaluate the knee adduction moment magnitude. PCA has also been shown to have limited sensitivity

to different biomechanical models used to define osteoarthritic gait patterns and PC1 is analogous to the knee adduction angular impulse (Newell et al., 2008). The objective of this study was to determine the immediate effect of a custom-fit valgus unloader brace on the overall magnitude of the net external knee adduction moment during gait in individuals with mild to moderate medial compartment knee OA.

3.2 Methods

3.2.1 Participants

Forty-one participants were recruited from local orthopedic assessment clinics primarily by the same orthopaedic surgeon (one participant was recruited by a second orthopaedic surgeon). All participants were prescribed a custom valgus unloader brace by the orthopaedic surgeon and showed clinical and radiographic evidence of mild to moderate medial compartment knee osteoarthritis (Altman et al., 1986). A subgroup of thirty-three participants voluntarily chose to purchase the valgus unloader brace and had the brace fit by an orthopaedic bracing specialist. No participants were candidates for total knee arthroplasty surgery at the time of testing. After initial recruitment, participants were contacted by phone and completed a screening questionnaire. The questionnaire included an assessment of function to verify that the participants had mild to moderate OA; all participants were able to walk a city block, jog 5 meters, and walk up and down stairs reciprocally (Hubley-Kozey et al., 2006). Participants had no other forms of arthritis, neuromuscular disease, cardiovascular disease, and no history of surgery to lower limb or orthopaedic pathology that would affect gait.

All participants wore a similar valgus unloader brace (Breg Fusion brace, $n = 32$; DonJoy brace, $n = 1$). A sample brace is pictured in Figure 3.1. Participants received one of three



Figure 3.1: Valgus unloader brace (Breg brand). The thigh and shank segments can be locked into an abducted position.

sizes of braces and were fit by the bracing specialist into 5° more abduction than their initial alignment. After approximately two weeks, the participants returned to the bracing specialist for a second fitting to ensure that they were wearing the brace correctly.

3.2.2 *Gait Analysis*

The present study was part of a larger study that required participants with a brace to visit the gait lab three times: upon recruitment, a second visit (approximately two weeks after the initial visit and after second fitting of brace by orthopaedic bracing specialist), and six months after the second visit. If participants did not receive a brace, they visited the lab for gait analysis only upon recruitment and after six months. Only the data from the second visit are compared between brace and no brace conditions in this study. Participants signed an informed consent form that was approved by the institutional review board regulations. Prior to gait analysis, demographic and anthropometric information was recorded including: age, mass, height, waist circumference at thinnest part of torso, thigh circumference at widest part of upper leg, shank circumference at widest part of lower leg, shoe size, and foot width at ball of foot. Participants graded their pain, stiffness, and function according to the Western Ontario and McMaster Universities Arthritis Index (WOMAC) (Bellamy et al., 1988). Gait analysis occurred after approximately two weeks of limited brace wear and after the secondary brace fitting and instruction by an orthopaedic brace specialist. This was intended to limit learned gait adaptations with brace wear, but allow for a correct fit. During testing, participants were not assisted or instructed in brace placement or fitting. Each participant was given a randomized condition order and brace and no brace gait trials were recorded.

For both the brace and no brace conditions, three-dimensional motion and force data of the affected side was collected during self-selected walking speeds using two 3020 Optotrak™ camera units (Northern Digital Inc., Waterloo, ON) and an AMTI force platform embedded in the walkway as previously described (Asthephen Wilson et al., 2011; Landry et al., 2007). Four triads of markers were placed on rigid bodies at the pelvis, thigh, shank, and foot of the affected side, and three markers were placed on the lateral malleolus, greater trochanter, and shoulder. Other anatomical landmarks were digitally identified during standing calibration. These included the right and left anterior and superior iliac spines, lateral epicondyle, medial epicondyle, fibular head, tibia tubercle, medial malleolus, second metatarsal, and heel. Marker motion was captured at 100 Hz and ground reaction

forces and moments were collected at 2000 Hz. Electromyography was also collected simultaneously using standardized procedures (Hubley-Kozey et al., 2006) at eight muscle sites: lateral and medial gastrocnemius, lateral and medial hamstring, vastus lateralis and medialis, rectus femoris, and gluteus medius. However, only motion and force data was analyzed in the current study.

Participants practiced walking to ensure a comfortable, self-selected velocity was reached. A group of five trials was performed wearing the brace and five without the brace. Trials were recorded if only the foot of the affected leg came into full contact with the force platform and gait velocity was within 10% of the self-selected walking velocity.

3.2.3 *Data Processing*

All data processing was performed using custom developed software written in Matlab (Landry et al., 2007). 3D motion and force data were digitally filtered at 8 Hz and 60 Hz, respectively, using a second order bidirectional Butterworth filter. The net external knee adduction moment (KAM) was calculated using an inverse dynamics model (Landry et al., 2007; Costigan et al., 1992) with direction defined by the floating axis of the joint coordinate system (Grood and Suntay, 1983). Moment waveforms were time normalized to 100% of gait cycle (from initial foot-strike of the affected leg on the force platform to the next foot-strike of the same leg) and the five trials for each participant were averaged. Moments were also normalized to body mass (Aststephen et al., 2008; Landry et al., 2007).

3.2.4 *Principal Component Analysis (PCA)*

Principal component analysis (PCA) was applied to the time normalized net external knee adduction moment waveforms. PCA is a pattern recognition technique that can reduce gait waveform data for effective interpretation (Aststephen et al., 2008; Landry et al., 2007). For the knee adduction moment waveform, a 137 x 101 data matrix \mathbf{X} was created that included the gait cycle data for all participants and all three laboratory sessions. The rows represented observations of the respective gait measure for participants during both brace and no brace conditions and some or all of the gait lab visits (depending on the participants' progress throughout the study), as well as participants who were prescribed a brace, but did not receive one for voluntary reasons (initial n = 17, second visit n = 74, six month n = 46, total n = 137). This large data set of waveforms was used to generate more stable principal components. Columns represented the 101 time points of the gait cycle (0 to 100%). The

first principal component was extracted from the covariance matrix of \mathbf{X} and this described the primary pattern of variability in the knee adduction moment waveform data, typically the overall magnitude of the knee adduction moment during the stance phase (Hatfield et al., 2010; Astephen Wilson et al., 2011; Landry et al., 2007). This PC score has been shown to be correlated to the area under the curve (Newell et al., 2008). Eigenvalues were also extracted from the covariance matrix of \mathbf{X} to describe the variation explained by each time point in the PC. A PC1-score was then calculated for each participant in the current study ($n = 33$) and for each condition (brace, no brace, 66 scores). The PC1-score was the projection of the participants' adduction moment data onto PC1, calculated as $\text{PC1-score} = (\mathbf{X})U_1$, where U_1 is the first eigenvector of the covariance matrix of the adduction moment data (Landry et al., 2007). An individual's PC1-score corresponded to the magnitude of a participant's external knee adduction moment waveform over stance phase.

3.2.5 *Brace vs. No Brace Analysis: Knee Adduction Moment Magnitude*

Since the purpose of the valgus unloader brace is to reduce medial compartment loads, the magnitude of the knee adduction moment was investigated due to its high correlation with medial compartment loads (Zhao et al., 2007; Erhart et al., 2010). Paired t-tests were used to determine condition (brace vs no-brace) differences in the knee adduction moment PC1-scores and in average walking speeds ($\alpha = 0.05$).

3.3 Results

3.3.1 *Brace vs. No Brace Analysis: Knee Adduction Moment Magnitude*

The participant demographic and anthropometric information, WOMAC scores, and speeds are presented in Table 3.1. There was no statistically significant difference in gait speed between the brace and no brace condition ($p = 0.299$). The first PC of the knee adduction moment described 53.7% of the variance in the adduction moment between participants. The PC1 eigenvector for the knee adduction moment is represented in Figure 3.2 with a shaded region showing the plot of the explained variation by PC1 (Astephen and Deluzio, 2005). There was no statistically significant difference in knee adduction moment magnitude (PC1 score) between the brace and no brace condition ($p = 0.342$) (Figure 3.3).

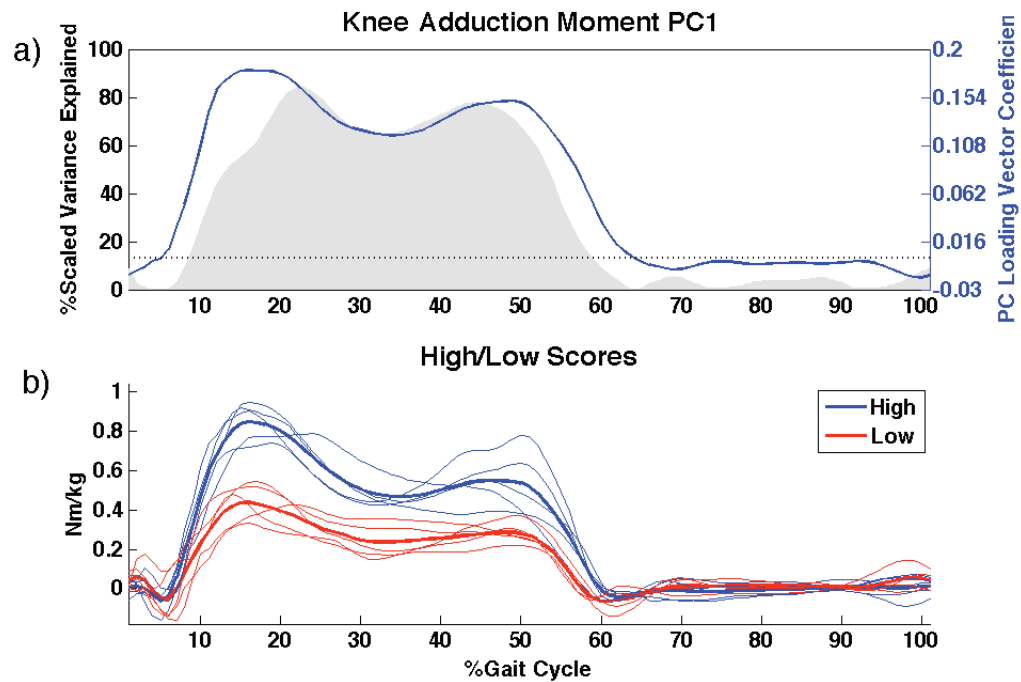


Figure 3.2: Knee adduction moment PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores. Blue lines are KAM waveforms with high scores, red lines are KAM waveforms with low scores, thin lines are the top or bottom fifth percentile of KAM waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms.

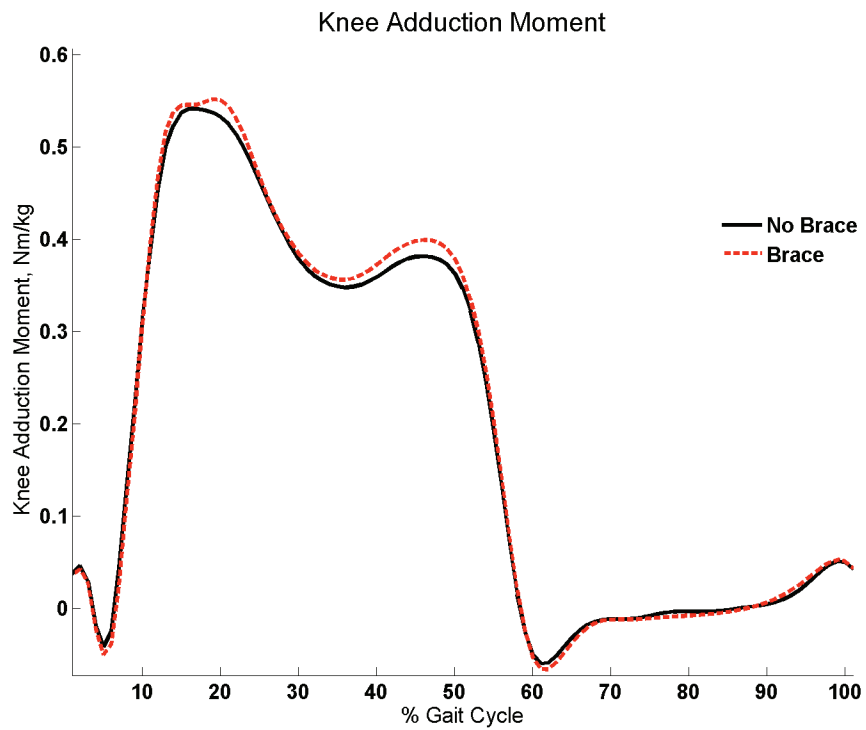


Figure 3.3: Knee adduction moment ensemble average waveform for all participants with a valgus unloader brace for the second visit only plotted by condition over the entire gait cycle. There was no significant difference between conditions for PC1 ($p = 0.342$).

Table 3.1: Participant Demographics. Data displayed as mean (standard deviation), where applicable.

	All Participants
Age (years)	57.6 (8.7)
Number	33
No. Males	25
Height (m)	1.73 (0.08)
Mass (kg)	95.6 (19.7)
BMI (kg/m ²)	31.6 (4.9)
Thigh Circumference (m)	0.54 (0.05)
Calf Circumference (m)	0.39 (0.04)
WOMAC Pain (/20)	6.3 (2.7)
WOMAC Stiffness (/8)	3.7 (1.6)
WOMAC Function (/68)	19.7 (9.7)
WOMAC Total (/96)	29.7 (12.3)
Speed: No Brace (m/s)	1.23 (0.15)
Speed: Brace (m/s)	1.24 (0.16)

3.4 Discussion

Because of the controversy in the literature, the purpose of this study was to evaluate if the valgus unloader brace altered the overall magnitude of the external knee adduction moment in participants with mild to moderate medial compartment knee osteoarthritis using Principal Component Analysis. After at least two weeks of brace wear, the external knee adduction moment magnitude for the participant group as a whole was not reduced during the brace condition when compared to the no brace condition. Although PC1 was used as a measure of magnitude instead of peak values, the results of the current study reflected that of previous studies that showed no significant change in the peak external knee adduction moment magnitude with brace wear (Gaasbeek et al., 2007; Pollo et al., 2002). In contrast, two studies did show a reduction in the peak external knee adduction moment with valgus unloader bracing (Self et al., 2000; Lindenfeld et al., 1997). However, Self et al. (2000) did not measure gait speed and spoke to a possible reduction in speed during the brace condition as a reason for a decrease in magnitude of the external knee adduction moment. The knee adduction moment magnitude is affected by gait speed (Landry et al., 2007). Lindenfeld et al. (1997) did not find differences in speeds, but the eleven participants had all recently undergone an arthroscopic surgery that would have

affected gait. It is possible that the small sample size had too much variability to detect a difference between speeds. Fantini Pagani et al. (2010a) found a reduction in the knee adduction angular impulse over stance phase with brace application, but the brace used in that study is not comparable to the braces used in this study as the thigh and shank segments extended almost the full length of the limb segments. Furthermore, that brace design was not typical of those prescribed for medial compartment knee OA. Given that the knee adduction moment PC1 score did not show a statistically significant change between conditions in this study, we can conclude that there was no systematic reduction in the overall knee adduction moment magnitude from the immediate application of the valgus unloader braces used in this study.

While the knee adduction moment has been correlated with medial joint contact force (Erhart et al., 2010; Zhao et al., 2007) and has been used extensively to evaluate the medial compartment loading (Andriacchi et al., 2004; Schipplein and Andriacchi, 1991), there are important limitations to consider in its interpretation as a surrogate measure of medial contact force. The net external knee adduction moment does not account for changes in the load distribution between contact forces and passive and active structures in the joint between conditions (Crowninshield and Brand, 1981). The external knee moment calculated by inverse dynamics is the sum of all external moments acting on the knee and is equated to all of the internal moments caused by the passive and active structures surrounding the knee joint. The problem of distributing this moment to individual structures in the joint to calculate individual structure forces is indeterminate as per the distribution problem described by Crowninshield and Brand (1981) (Figure 3.4). Without calculating all of the contact loads and forces within the joint, as is done in a forward dynamics analysis, there are not enough known variables to calculate medial compartment force. Muscles, which can increase peak force felt by the knee by 3-5 times body weight (Taylor et al., 2004), may be a particularly important factor to the loading distribution as changes in agonist-antagonist co-activation have been found with valgus unloader brace wear (Ramsey et al., 2007). The addition of another structure that can resist the knee adduction moment, such as the brace, could account for some reduction in medial joint force that is not measured in the inverse dynamics analysis, but it is only one of many structures over which the load is distributed (Figure 3.4). As well, the brace abduction moment has been shown to be small when used on an osteoarthritic population. Using an instrumented

brace, Pollo et al. (2002) showed that the abduction moment applied by the brace was no more than 10% of the net external knee adduction moment magnitudes found in this study. However, even though the brace abduction moment magnitude is likely small, there could still be large changes in muscle forces between the brace and no brace conditions which need to be evaluated prior to conclusions made about changes in the medial compartment contact force.

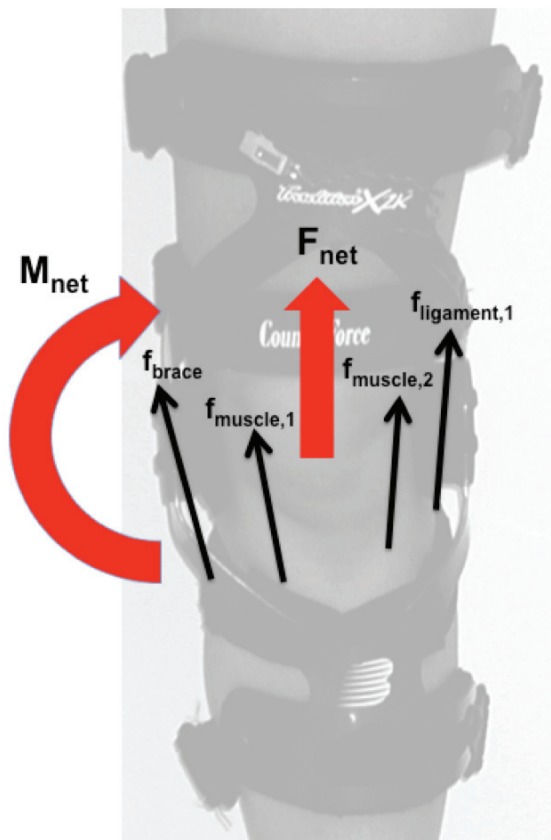


Figure 3.4: The forces and moments acting on the shank from the passive and active structures around the knee joint sum to the net external force and moment (F^{net} , M^{net}). The brace is only one of many structures that are distributed the overall load, and in the case of muscular co-contraction, sometimes in opposing directions.

As the knee joint moves and carries loads in three planes, assessing only the knee adduction moment magnitude limits the understanding of the knee joint mechanics and how they change with brace application. Pollo et al. (2002) incorporated flexion and extension moments in his analytical model and found a reduction in average medial compartment contact force, suggesting that moments other than those in the frontal plane may be affected by brace wear. Thus, it is important to study kinetic as well as kinematic changes in all three anatomical planes to better understand the method by which the valgus unloader brace effects change during gait.

Although there are limitations in the knee adduction moment calculated with inverse dynamics, it is a commonly accepted indicator of risk of knee OA progression and a key piece to the mechanics of the knee joint. Therefore, it is still an important measure that needs to be quantified to better understand the biomechanical effects of valgus unloader brace application and its ability to slow disease progression.

It is likely that a valgus unloader brace may have different effects on the external knee adduction moment for some participants than others and all participants may not exhibit the same gait adaptations caused by a valgus unloader brace. This could result in groups of participants who display different changes in the external knee adduction moment due to brace wear, which may nullify any overall brace effect. The next chapter will explore this concept.

3.4.1 Conclusion

The results of this study did not indicate that the knee adduction moment was systematically reduced with valgus unloader brace wear for the entire participant population. It may indicate that the response of all participants to the application of the brace was not homogeneous. Other changes in gait mechanics need to be investigated to understand the effects on the brace on gait mechanics in all three planes.

CHAPTER 4

DOES THE BRACE AFFECT ALL PARTICIPANTS IN THE SAME WAY?

4.1 Introduction

Valgus unloader braces are prescribed to patients with mild to moderate medial compartment knee osteoarthritis (OA) to relieve pain and increase stability (Zhang et al., 2008). They are designed to reduce the medial compartment forces associated with OA through application of an abduction moment that counters adduction loads (Pollo et al., 2002). The external knee adduction moment magnitude, which is the net effect of these loads, has been associated with medial compartment force in the knee and therefore, is often used as a surrogate measure of medial compartment contact force (Erhart et al., 2010; Zhao et al., 2007).

The effects of valgus unloader bracing on the net external knee adduction moment have been examined in other studies (Fantini Pagani et al., 2010b; Gaasbeek et al., 2007; Pollo et al., 2002; Self et al., 2000; Lindefeld et al., 1997); however, there is debate as to the effects on the overall magnitude. Gaasbeek et al. (2007) and Pollo et al. (2002) did not find a reduction in peak external knee adduction moment, while Lindefeld et al. (1997), Self et al. (2000), and Fantini Pagani et al. (2010b) found significant reductions in either first or second peaks. Chapter 3 found no statistically significant reduction in the overall stance external knee adduction moment magnitude on the participant group as a whole. The discrepancies in results between these studies can be explained by changes in walking velocity between brace and no brace conditions, small sample sizes, differences between

the biomechanical model used for analysis, and in some cases bracing conditions that do not represent typical brace prescription (Fantini Pagani et al., 2010a; Gaasbeek et al., 2007; Pollo et al., 2002; Self et al., 2000; Lindenfeld et al., 1997). An alternative explanation is that the valgus unloader brace does not have a consistent effect on all patients given the heterogeneity within the OA population. In preliminary findings of Otis et al. (1996), it was found that participants exhibited varying changes in medial joint load with brace wear, including two of six participants who had increased medial joint load. The sample size in this study was small, but the results do raise the question whether subgroups exist within an OA sample that would respond differently to brace wear.

To expand our understanding of the effect of valgus unloader brace application during gait, it is important to measure the changes in kinetics and kinematics of the knee joint on all three planes. The presence and severity of osteoarthritis have been shown to effect 3D loading of the knee (Aststephen et al., 2008; Landry et al., 2007; Gok et al., 2002; Baliunas et al., 2002; Hurwitz et al., 2002; Kaufman et al., 2001; Sharma et al., 1998) and these changes may be indicators of progression. However, very few of these studies were longitudinal in design and the limits of these comparisons must be considered when evaluating possible biomechanical factors of progression. The peak knee adduction moment magnitude has been associated with progression through a longitudinal study (Miyazaki et al., 2002). Increased peaks and magnitude of the adduction moment have been linked with the presence of OA (Gok et al., 2002; Hurwitz et al., 2002; Landry et al., 2007) and increases in the mid-stance values have been related to severity (Aststephen et al., 2008). In the sagittal plane, a reduced knee flexion moment magnitude in early stance (Aststephen et al., 2008; Landry et al., 2007; Kaufman et al., 2001) and a reduced extension moment in late stance (Gok et al., 2002; Baliunas et al., 2002) have been shown in an OA population. In one study, participants with severe OA experienced a knee flexion moment throughout the entire stance phase, without an extension moment at mid-late stance (Aststephen et al., 2008). There is disagreement in the literature as to the effects of OA on the knee rotation moment (Landry et al., 2007; Gok et al., 2002; Kaufman et al., 2001); however, this may be due to differences between the waveform analysis techniques used in these studies as changes in timing and magnitude of the knee rotation moment have been shown with increasing OA severity (Aststephen et al., 2008).

Knee OA has also been associated with changes in 3D motion of the knee joint (Kuroyanagi et al., 2011; Astephen et al., 2008; Landry et al., 2007; Chang et al., 2004; Gok et al., 2002; Kaufman et al., 2001). Moderate OA participants exhibit less flexion throughout the gait cycle than controls (Landry et al., 2007; Gok et al., 2002; Kaufman et al., 2001) and severe OA participants have been shown to have less extension in late stance phase (Astephen et al., 2008). In the frontal plane, the existence of varus thrust, an abrupt lateral motion of the knee in early stance during weight acceptance (Chang et al., 2004, 2010), has been associated with the progression of knee OA in a longitudinal study (Chang et al., 2004) and its magnitude has been linked with the magnitude of the knee adduction moment, an indicator of disease progression risk (Kuroyanagi et al., 2011). No studies have evaluated the effect of the knee rotation angle on knee OA progression. As more studies evaluate kinematic and kinetic changes with knee progression using a longitudinal design, other biomechanical indicators of the risk of knee OA progression may become apparent. A recent study by (Henriksen et al., 2011) showed that the knee rotation moment peak is associated with the knee adduction moment peak and suggests that the role of knee rotation moment in the disease process of knee OA should be investigated more thoroughly.

The brace has been shown to treat OA illness using self-reported measures of pain and function (Gaasbeek et al., 2007; Richards et al., 2005; Kirkley et al., 1999). Patients have also been shown to walk faster with and without the brace after extended periods of brace wear (Gaasbeek et al., 2007), although no change in gait speed with immediate application of the brace was seen in Chapter 3. Self et al. (2000) and Otis et al. (2000) argued that other kinetic and kinematic parameters more than just the abduction moment applied by the brace may be responsible for its clinical success. Pollo et al. (2002) used an analytical model that incorporated changes in sagittal plane variables with brace wear to determine that the average medial compartment contact force over stance phase reduced with brace wear. Several studies have examined 3D kinematic and kinetic variables other than the knee adduction moment; however this has been focused mainly on sagittal plane variables. The brace has been shown to reduce the range of flexion/extension motion (Richards et al., 2005; Gaasbeek et al., 2007; Davidson et al., 1998). One study showed an increase in flexion moment at early stance and decrease in the extension moment at late stance (Toriyama et al., 2011), although this was attributed to a small increase in gait speed

with brace application. Increased magnitudes of rotation angle and increased adduction at toe-off have been reported in the transverse plane (Davidson et al., 1998). In the frontal plane, condylar separation with valgus unloader brace wear at foot-strike has been shown (Komistek et al., 1999) and the brace reduced dynamic adduction angle range (frontal plane excursions toward knee adduction) during weight acceptance (Ramsey et al., 2007); however the peak knee adduction angle during the gait cycle did not change with brace wear, showing that the brace does not consistently effect the frontal plane alignment over the entire gait cycle.

Studies which have examined kinematic and kinetic variables other than the knee adduction moment have analyzed discrete variables such as waveform peaks, range of motion, or the magnitude at several time periods (Richards et al., 2005; Gaasbeek et al., 2007; Ramsey et al., 2007; Davidson et al., 1998). Although the use of discrete variables provides valuable important information about knee joint loading and motion, they provide limited information about the temporal features of gait waveforms consistent with the dynamic motion of gait. Gait variables have been shown to change in magnitude and shape over the entire gait cycle between participants with varying degrees of OA severity (Landry et al., 2007; Astephen et al., 2008). Principal Component Analysis (PCA) is a pattern recognition technique that has been shown to be effective at describing and comparing the temporal and magnitude aspects of gait waveforms over the entire gait cycle in asymptomatic and OA participants (Deluzio and Astephen, 2007; Astephen Wilson et al., 2011; Hatfield et al., 2010; Landry et al., 2007).

Based on the state of the literature on valgus unloader bracing, there is a need to determine if subgroups of participants who respond differently to brace application exist within the OA population. Secondly, to expand our understanding of the effects of bracing on gait mechanics, it is necessary to investigate the 3D kinematic and kinetic changes that occur with valgus unloader brace application. The objectives of this study were twofold: 1) to identify subgroups whose overall knee adduction moment magnitude i) decreased with brace application, ii) did not change with brace application, and iii) increased with brace application; 2) to determine differences in 3D knee kinematic and kinetic features after brace application and among subgroups who exhibit varying changes in net knee external adduction moment with the immediate application of a valgus unloader brace.

4.2 Methods

Similar recruitment and laboratory protocol to Chapter 3 were used.

4.2.1 Data Processing

All data processing was performed using custom developed software written in Matlab (Landry et al., 2007). 3D motion and force data were digitally filtered at 8 Hz and 60 Hz, respectively, using a second order bidirectional Butterworth filter. 3D joint angles at the knee were calculated according to the joint coordinate system described by Grood and Suntay (1983). Net external joint moments were calculated using an inverse dynamics model (Landry et al., 2007; Costigan et al., 1992) and directions defined using the joint coordinate system (Grood and Suntay, 1983). Knee angle and moment waveforms were time normalized to 100% of the gait cycle and the five trials for each participant were averaged. Because the brace has been shown to affect knee adduction excursions during weight acceptance, the knee adduction angle was only analyzed during stance phase and therefore, truncated to 0-60% of the gait cycle. All joint moments were normalized to body mass (Asthephen et al., 2008; Landry et al., 2007). Moment waveforms were truncated to one gait cycle from initial foot-strike of the affected leg on the force platform to the next foot-strike of the same leg.

4.2.2 Principal Component Analysis (PCA)

Principal component analysis (PCA) was applied to the time normalized knee adduction, flexion, and rotation angles and net knee external moment waveforms. PCA is a pattern recognition technique that can reduce gait waveform data for effective interpretation (Asthephen Wilson et al., 2011; Landry et al., 2007). For each waveform type, a 137 x 101 data matrix or 137 x 61 data matrix, in the case of knee adduction angle, \mathbf{X} , was created that included the gait cycle data for all participants and all three laboratory sessions. The rows represented observations of the respective gait measure for participants during both brace and no brace conditions and some or all of the gait lab visits (depending on the participants' progress throughout the study), as well as participants who were prescribed a brace, but did not receive one for voluntary reasons (initial n = 17, second visit n = 74, six month n = 46, total n = 137). Columns represented the 101 time points of the gait cycle (0 to 100%) or 61 in the case of the adduction angle. Eigenvectors, referred to as principal

components (PCs), were extracted from the covariance matrix of \mathbf{X} , and these described the primary patterns of variability in the waveform data set. The first three PCs were included in the statistical analysis if each contributed to at least 4% of the variance individually and cumulatively more than 80% of the total variance in the waveform measure. A PC-score for each PC was then calculated for each participant in the current study ($n = 33$), on each waveform, and for each condition (brace, no brace, $n = 66$). The PC-score was the projection of the participants data onto the PC, collectively calculated as $\text{PC-score} = (\mathbf{X})\mathbf{U}$, where \mathbf{U} is the matrix of the reduced number of eigenvectors (first three) (Landry et al., 2007). An individual's PC-score represented how closely an individual's waveform pattern corresponded to the pattern described by that PC. To better interpret the PCs and the meaning of the scores, the relative importance of the PC at each portion of the gait cycle is evaluated. The percent variation explained, *Percent Variation Explained*(i,j), is a measure of the amount of variation in the participant waveforms at each time point that is explained by the PC, PC_j , at that time point, i (Deluzio and Astephen, 2007).

4.2.3 Analysis of Groups

A primary goal of the brace is to reduce the medial joint forces at the knee and three groups were identified based on how the external knee adduction moment, which has been correlated to medial compartment contact force, changed with brace application. A difference in the knee adduction moment waveform PC1 scores (which represented the overall magnitude of the moment during stance) between the brace and no brace conditions was calculated for each participant. Three subgroups were identified based on this difference. The three subgroups included participants who 1) increased, 2) decreased, or 3) did not change their knee adduction moment PC1 score when the brace condition was compared to the no brace condition. The threshold for determining these subgroups was set at $\pm 10\%$ of the total range of percent differences in PC scores between conditions relative to the no brace condition among all participants. The threshold was calculated by:

$$\Delta\%PC = \frac{PC_{\text{brace}} - PC_{\text{no brace}}}{PC_{\text{no brace}}} \quad (4.1)$$

$$x_t = 0.1(\Delta\%PC_{\text{max}(+)} - \Delta\%PC_{\text{max}(-)}) \quad (4.2)$$

where $\Delta\%PC$ is the percent change in PC score for each participant, $\Delta\%PC_{\text{max}(+)}$ is the greatest positive percent change in PC score over all participants, $\Delta\%PC_{\text{max}(-)}$ is the

greatest negative percent change in PC score over all participants, and x_t is the threshold value.

The participants that did not change their knee adduction moment PC1 score by more than the 10% threshold change in PC score with brace wear were in the No Change group. The participants that increased their knee adduction moment PC1 score by more than the 10% threshold from the no brace condition to the brace condition were in the Increase group and the participants that decreased in PC1 score by more than the 10% threshold were in the Decrease group. Therefore,

$$\text{Increase} : \Delta\%PC > x_t \quad (4.3)$$

$$\text{No Change} : -x_t < \Delta\%PC < x_t \quad (4.4)$$

$$\text{Decrease} : \Delta\%PC < -x_t. \quad (4.5)$$

4.2.4 Varus Thrust

In addition to the PC scores, three discrete variables were calculated for the frontal plane angle based on clinical observations previously reported (Chang et al., 2010, 2004). For each participant, a maximum and minimum angle in early stance (0-30% of the gait cycle) was selected from the knee adduction angle waveform and the difference between these values was considered the varus thrust (Figure 4.1, solid line). Each participant's waveform was reviewed to confirm that these values captured the beginning and end of an 'abrupt lateral motion,' which was defined in this study as a change of more than 0.5° over less than 10% of the gait cycle between 0-30% of the gait cycle. For participants that exhibited an abrupt change in ab/adduction angle but that did not have a distinct minimum or maximum angle before and after the thrust, the minimum and maximum were chosen as the point at which the slope of the varus thrust decreased by greater than 50% (Figure 4.1, dotted line). Participants that did not exhibit varus thrust (an abrupt change in adduction of more than 0.5°) were removed from the analysis. Figure 4.1 is a pictorial representation of the determination of minimum and maximum angle and varus thrust.

4.2.5 Statistical Analysis

A one-way Analysis of Variance was used to determine group main effects in the means and standard deviations of demographic and anthropometric data. A two-way Analysis of

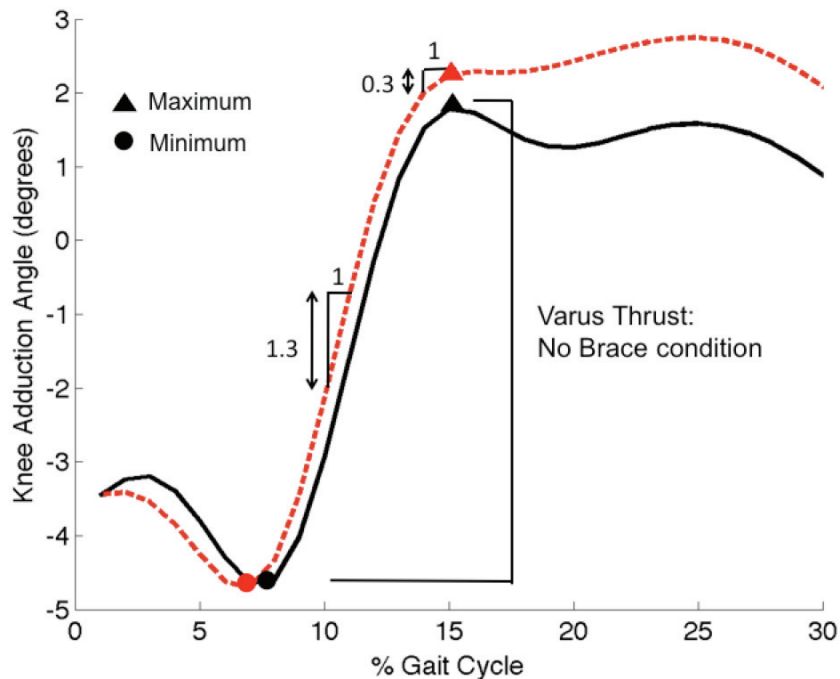


Figure 4.1: Example of varus thrust calculation. Knee adduction angle waveform for 0-30% of the gait cycle for one participant during both conditions. The brace condition is the red dashed line and the no brace condition is the black solid line. For this participant, it was necessary to use both methods of determining varus thrust for the different conditions. The no brace condition had a minimum and maximum in the adduction angle waveform before and after an 'abrupt lateral motion' or change in adduction angle between 6 and 15% gait cycle. The minimum and maximum angle were represented by the circle and triangle markers, and the no brace varus thrust is represented by the black bar. During the brace condition, there was no apparent peak in knee adduction angle after the abrupt change in alignment. The maximum angle was chosen as the point at which the slope became at least 50% less than the slope during the abrupt change. The slopes are noted on the diagram during the abrupt change ($1.3^{\circ}/\% \text{gait cycle}$) and at the end of varus thrust ($0.3^{\circ}/\% \text{gait cycle}$).

Table 4.1: Descriptive data for all participants and by group. Data is presented as means (standard deviation).

	Increase Group	No Change Group	Decrease Group
Age (years)	56.8 (8.4)	57.4 (9.3)	58.9 (9.4)
Number	13	10	10
Males:Females	11:2	8:2	6:4
Height (m)	1.74 (0.067)	1.74 (0.087)	1.72 (0.11)
Mass (kg)	98.8 (18.5)	95.7 (20.6)	91.5 (21.5)
BMI (kg/m ²)	32.4 (4.5)	31.6 (5.2)	30.7 (5.5)
Speed: No Brace (m/s)	1.21 (0.13)	1.28 (0.18)	1.21 (0.15)
Speed: Brace (m/s)	1.21 (0.13)	1.28 (0.18)	1.23 (0.19)
WOMAC Pain	6.5 (2.6)	6.4 (2.4)	5.9 (3.2)
WOMAC Stiffness	4.1 (2.1)	3.5 (1.3)	3.5 (1.3)
WOMAC Function	17.0 (11.4)	21.1 (8.5)	21.7 (8.4)
WOMAC Total	27.6 (14.3)	31.0 (10.9)	31.1 (11.7)

Variance was used to determine group and condition main effects and group by condition interactions in angle and moment PC-scores, adduction angle maxima, adduction angle minima, and varus thrust magnitude ($\alpha = 0.05$). Significant differences were further analyzed using post-hoc Tukey corrected t-tests.

4.3 Results

4.3.1 Participants

4.3.1.1 Groups

The group demographic data, speeds, and WOMAC scores are represented in Table 4.1. There were no statistically significant differences ($p > 0.05$) between groups for any of these measures.

The knee adduction moment waveforms for each group are depicted in separate graphs in Figure 4.2. For each group, an increase, a decrease, or no change in the overall magnitude of the adduction moment between the brace condition and no brace condition is apparent. As expected, there was a significant group by condition interaction for PC1 score ($p < 0.001$). The Increase group had a significantly larger knee adduction magnitude during the brace condition than the no brace condition ($p < 0.001$) and the Decrease group had

a significantly smaller knee adduction moment magnitude during the brace condition than the no brace condition ($p < 0.001$). This supports that the groups did have different responses based on the classification procedure employed. The No Change group did not have statistically significant difference between conditions for knee adduction moment PC1 score ($p = 1.00$). The No Change group had a higher knee adduction moment magnitude for both conditions than the other two groups ($p < 0.001$).

4.3.2 *Two-Way ANOVA*

A description of each loading vector and the results of the two-way ANOVA on the principal components and varus thrust are shown in Table 4.2. There were six significant condition effects and four significant interactions. Significant differences are discussed in the following sections.

4.3.2.1 *Condition Effects*

The condition effects are summarized in Table 4.3. The no brace condition had a higher extension moment before toe off relative to the moment at midstance (PC3) (Figure 4.3a). In the transverse plane, the knee was more externally rotated at midstance during the brace condition (PC3) (Figure 4.3b). In the frontal plane, the knee had a larger range of ab/adduction motion between early and late stance (PC2) and a larger range of ab/adduction motion between 5-20% of the gait cycle (PC3) (Figure 4.3c). The latter effect, PC3 (Figure 4.3d) represented 6.4% of the variance among knee adduction angle waveforms in the PCA analysis and was correlated with the varus thrust measure ($r^2 = 0.49$, $p < 0.001$). The varus thrust magnitude was significantly larger for the brace condition than the no brace condition as depicted in (Figure 4.3c). Table 4.4 shows the means and standard deviations of the minimum and maximum adduction angles in early stance and magnitudes of varus thrust for each condition. Two participants (one from each of the Decrease and Increase groups) were removed from the analysis as they did not exhibit varus thrust.

4.3.2.2 *Group Effects*

There were no significant group effects ($p > 0.05$) although there was a trend toward differences among groups for the knee rotation moment PC3 ($p = 0.05$). PC3 represented 8.9% of the variability in the knee rotation moment PCA analysis. The loading vector represented an increased magnitude of internal rotation moment in early stance and the beginning of swing (Figure 4.4b). The Increase group exhibited this pattern more than any other group, followed by the Decrease and No Change group, respectively (Figure 4.4a).

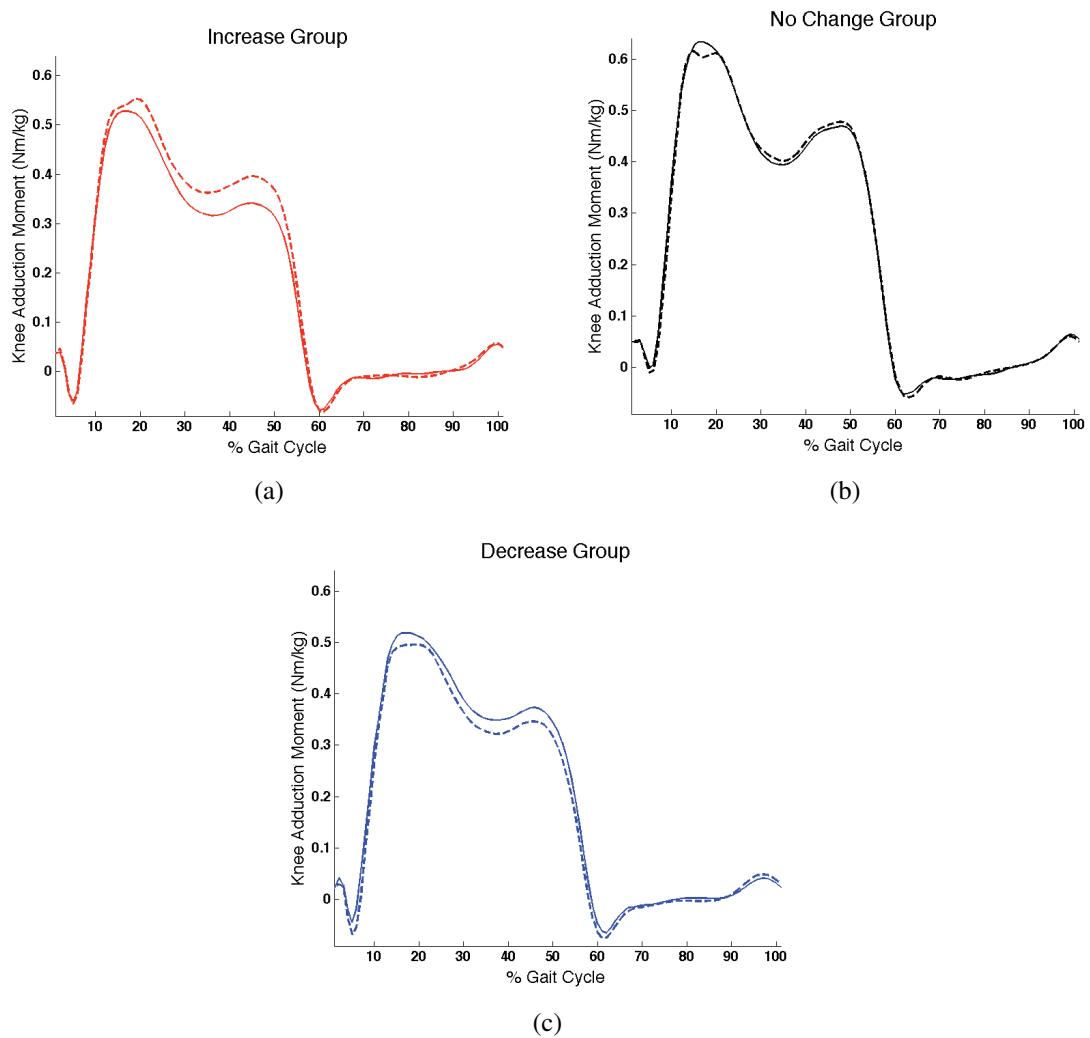


Figure 4.2: Mean knee adduction moment waveforms of the (a) Increase, (b) No Change, and (c) Decrease groups for both conditions over 100% of the gait cycle. The brace condition and no brace condition are represented by dotted and solid lines, respectively.

Table 4.2: Results of two-way ANOVA on loading vectors and varus thrust. P-values are listed for condition and group main effects, as well as condition and group interactions. A brief description of the interpretation of each loading vector is included (with % variation explained in brackets).

Measure	PC	Description (% Variance Explained)	Group	Condition	Interaction
Knee Adduction Moment	1	Magnitude during stance (53.7%)	0.12	0.34	< 0.001
	2	Relative magnitude of first and second abduction peaks (24.6%)	0.68	0.10	0.14
	3	Relative magnitude of first two abduction peaks and valley at midstance (6.4%)	0.19	0.75	0.12
Knee Flexion Moment	1	Magnitude (56.7%)	0.16	0.56	0.10
	2	Relative magnitude of extension during midstance and flexion at toe-off (28.7%)	0.76	0.07	0.75
	3	Range of extension to flexion moment at early-mid stance and flexion to extension moment at mid-late stance (4.9%)	0.51	0.01	0.13
Knee Rotation Moment	1	Magnitude (50.0%)	0.04	0.52	0.01
	2	Relative magnitude of external rotation at early stance and internal rotation at late stance (25.4%)	0.23	0.85	0.10
	3	Internal rotation at heel strike and internal rotation during swing phase (8.9%)	0.05	0.40	0.80
Knee Adduction Angle	1	Magnitude (65.7%)	0.63	0.25	0.36
	2	Overall adduction to abduction change at 35% of the gait cycle (18.0%)	0.18	0.01	0.21
	3	Abduction to adduction change during initial stance phase (8.4%)	0.73	0.03	0.63
Knee Flexion Angle	1	Magnitude (57.3%)	0.14	0.10	0.02
	2	Represents a phase shift in timing of flexion peak during swing phase (21.3%)	0.60	0.02	0.47
	3	Relative magnitude of flexion at early stance and extension motion in late stance (7.4%)	0.69	0.35	0.37
Knee Rotation Angle	1	Magnitude (61.4%)	0.81	0.50	0.41
	2	Relative magnitude of rotation angle at early stance, late stance, and late swing (17.4%)	0.40	<0.001	0.03
	3	Relative magnitude of rotation angle at early stance, early swing, and late swing (10.1%)	0.64	< 0.001	0.53
Varus Thrust			0.90	0.001	0.15
Minimum Adduction Angle			0.60	0.06	0.22
Maximum Adduction Angle			0.72	0.07	0.09

Table 4.3: Results of the post-hoc analysis on condition effects.

Variable	Post-hoc description
Knee Flexion Moment PC3	No brace condition had higher extension moment before toe off relative to moment at midstance
Knee Adduction Angle PC2	Brace condition had larger range of motion between early and late stance
Knee Adduction Angle PC3	Brace condition had larger range of motion between 5 and 20%
Knee Flexion Angle PC2	Brace condition had earlier flexion peak during swing
Knee Rotation Angle PC3	Brace condition was more externally rotated at midstance, but similar rotation angle during swing
Varus Thrust Magnitude	Brace condition had larger varus thrust magnitude

Table 4.4: Varus thrust magnitude and maxima and minima of the knee adduction angle in early stance by condition. Data is displayed as mean (standard deviation). The varus thrust magnitude was larger for the brace condition than the no brace condition. Significant differences are highlighted with * (n = 31).

	No Brace	Brace
Varus Thrust (°)*	3.63 (2.2)	4.18 (2.0)
Minimum Angle (°)	-0.705 (2.1)	-0.882 (2.2)
Maximum Angle (°)	2.92 (1.9)	3.30 (1.9)

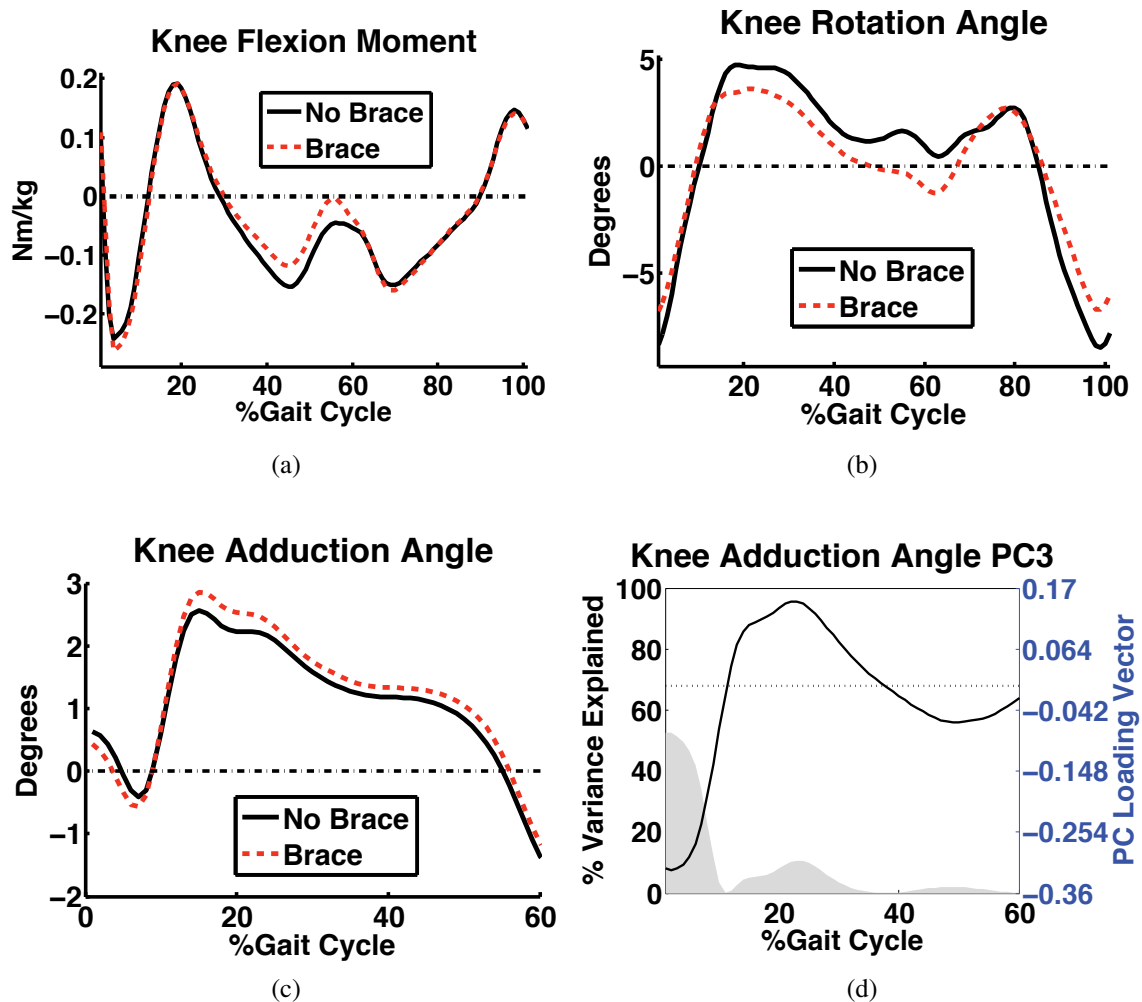


Figure 4.3: Condition effects. a) Mean knee flexion moments by condition ((+) flexion, (-) extension) b) mean knee rotation angles by condition ((+) internal rotation of tibia, (-) external rotation of tibia) c) mean knee adduction angles by condition ((+) is adduction, (-) is abduction) and d) knee adduction angle PC3 loading vector. The no brace condition had a higher extension moment before toe off relative to the flexion moment at midstance ($p = 0.010$) (PC3) (a). In the rotation angle, the brace condition is more externally rotated at midstance, but returns to a similar rotation angle during swing as the no brace condition ($p < 0.001$) (PC3) (b) ((+) internal rotation of tibia, (-) external rotation of tibia). The conditions were significantly different for knee adduction angle PC2 and PC3 ($p = 0.011$ and $p = 0.029$). The brace condition had increased range of motion between early and late stance (PC2) (c) as well as increased range between 5-20% (PC3 loading vector) (c,d). The brace condition had an increased varus thrust magnitude than the no brace condition ($p = 0.001$) (c).

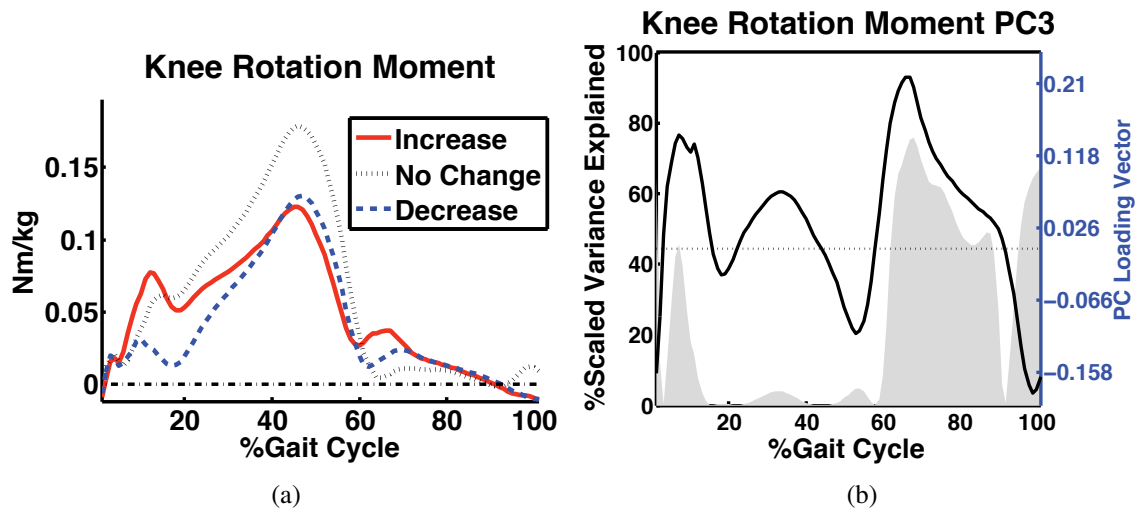
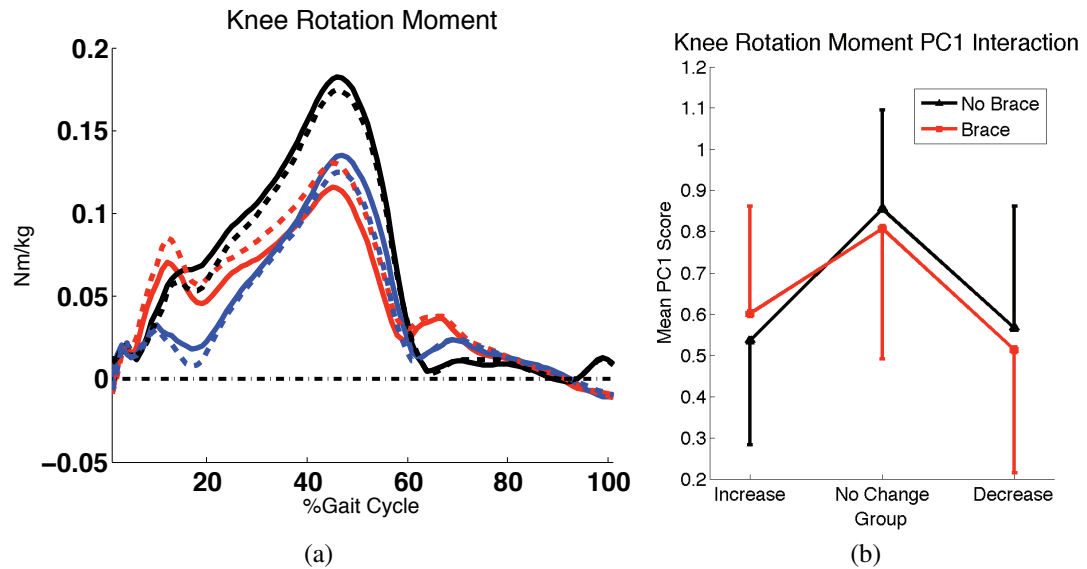


Figure 4.4: Group trend. Mean knee rotation moments by group (a) ((+) internal rotation acting on tibia, (-) external rotation moment acting on tibia). There was a trend toward a group effect for the knee rotation moment PC3, representing an increase in the internal rotation moment at early stance and early swing (b) (8.9% of variability in PCA analysis). In order from highest to lowest PC3 score, the groups were: Increase, Decrease, No Change.

4.3.2.3 Interactions

The knee rotation moment and knee flexion angle had significant condition by group interactions in the overall magnitude of the waveforms (PC1, $p = 0.01$, $p = 0.02$). For the rotation moment, the No Change group had a larger stance phase moment than both other groups for both conditions (PC1, $p < 0.0001$). During the brace condition, the Increase group exhibited a trend toward larger rotation moments than the Decrease group (PC1, $p = 0.06$) (Figure 4.5). The Decrease group had increased overall flexion angles during the gait cycle than the other two groups for both conditions ($p < 0.0001$) and the Increase group had slightly decreased flexion during the brace condition compared to the no brace condition (PC1, $p = 0.015$) (Figure 4.6).

There was a significant condition by group interaction for the knee rotation angle PC2 ($p = 0.03$), which represented a difference operator between rotation angles at early stance and early swing (high score was a higher relative difference in rotation angle between early stance and early swing) (Figures 4.7a and 4.7b). During the no brace condition, the No Change group had greater change from external to internal rotation from early stance to early swing than the other two groups ($p = 0.006$) (Figure 4.7). During the brace

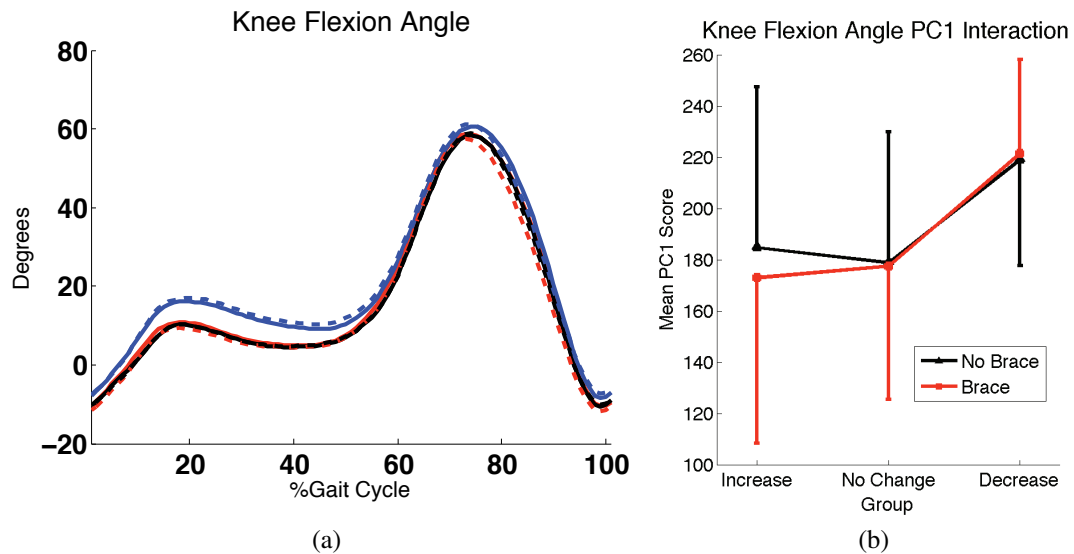


Between Group Differences

Between Condition Differences	No Brace	Increase	<	No Change	>	Decrease	=	Increase
		=		=		=		=
	Brace	Increase	<	No Change	>	Decrease	<	Increase

(c)

Figure 4.5: Mean knee rotation moments by group (Increase: red, No Change: black, Decrease: blue) and condition (Brace: dotted, No Brace: solid) are expressed in (a) ((+) internal rotation moment acting on tibia, (-) external rotation moment acting on tibia). There was a significant condition by group interaction for PC1 representing the magnitude of the moment ($p = 0.01$). A description of the post-hoc analysis is shown in c). A "<" or ">" indicates that the group and condition was significantly less than or greater than the next group and condition. A "=" indicates that there was no significant difference. The No Change group had a higher magnitude than both groups for both conditions. During the brace condition, there was trend for the Increase group to have higher magnitudes than the Decrease group ($p = 0.06$).



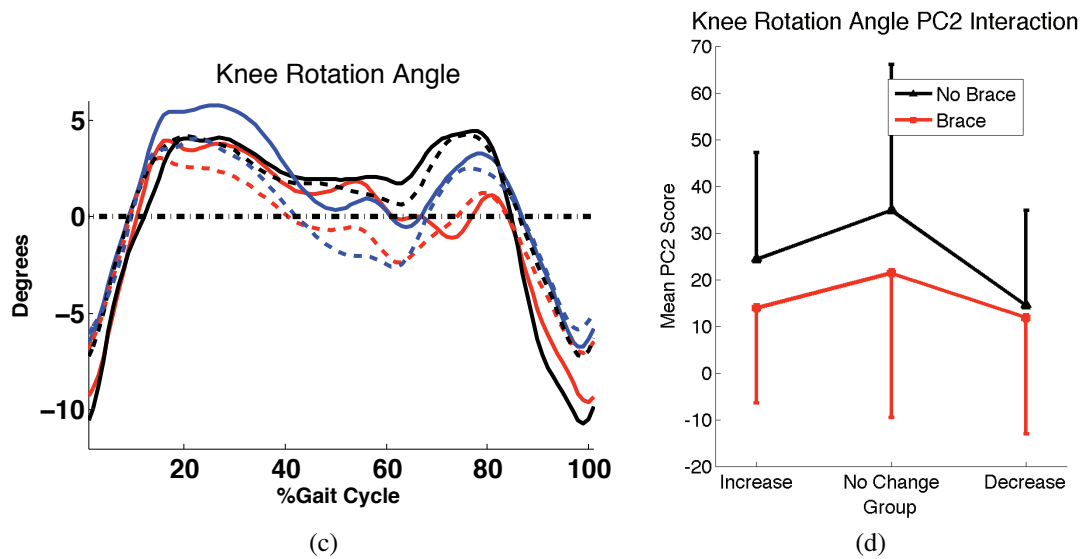
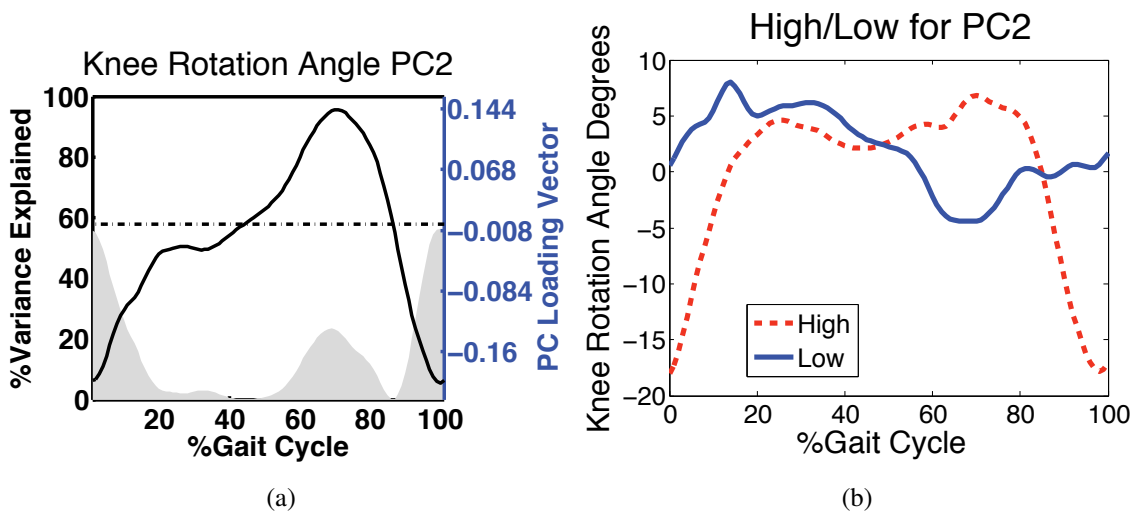
(a)

(b)

		Between Group Differences						
Between Condition Differences	No Brace	Increase	=	No Change	<	Decrease	>	Increase
		>		=		=		>
	Brace	Increase	=	No Change	<	Decrease	>	Increase

(c)

Figure 4.6: Mean knee flexion angle by group (Increase: red, No Change: black, Decrease: blue) and condition (Brace: dotted, No Brace: solid) are expressed in (a) ((+) flexion, (-) extension). There was a significant condition by group interaction for PC1 representing the magnitude of the flexion angle ($p = 0.02$). A description of the post-hoc analysis is shown in c), which uses the same legend described in Figure 4.5. The Decrease group had a higher magnitude than both groups for both conditions. During the brace condition, the Increase group had a lower flexion angle magnitude than during the no brace condition.



		Between Group Differences						
Between Condition Differences	No Brace	Increase	<	No Change	>	Decrease	=	Increase
		>		>		=		>
	Brace	Increase	=	No Change	>	Decrease	=	Increase

(e)

Figure 4.7: There was a significant condition by group interaction for knee rotation angle PC2 (a) representing a range between rotation angles at early stance and early swing (a,b)(p = 0.03). A description of the post-hoc analysis is shown in d), which uses the same legend described in Figure 4.5. During the no brace condition, the No Change group had a greater change from external to internal rotation from early stance to early swing when compared to the other groups (c,d,e) ((+) internal rotation of tibia, (-) external rotation of tibia). During the brace condition, the No Change group had a greater range than the Decrease group (c,d,e). The No Change and Increase group had a greater change in rotation angle from early stance to early swing for the no brace condition than the brace condition (c,d,e).

condition, the No Change group had greater change from external to internal rotation from early stance to early swing than the Decrease group ($p = 0.024$). Both the No Change and Increase group had a greater change in rotation angle from early stance to early swing for the no brace condition than the brace condition, moving to less rotation range with brace on than off ($p < 0.001$ and $p = 0.032$).

4.4 Discussion

There existed three groups of participants that could be separated based on the change in external knee adduction moment magnitude caused by applying a valgus unloader brace. This shows that the brace did not have a consistent effect on the knee adduction moment of osteoarthritic knees. Walking velocity has been shown to affect the knee adduction moment magnitude (Landry et al., 2007); however, there were no differences in walking velocity between groups or conditions. Therefore, the differences between participants in the change in knee adduction moment magnitude is evidence that not all participants incur the same alterations to knee joint loading with brace wear. Although the knee adduction moment has been correlated with medial joint load, this correlation is limited within this analysis as it was performed without brace application. The knee adduction moment calculated with inverse dynamics provides a limited estimation of knee force distribution (Crowninshield and Brand, 1981) as it gives no indication of increased compartment loads due to agonist-antagonist muscle forces. Co-activation of musculature around the knee has been shown to exist within the OA population (Hubley-Kozey et al., 2008; Childs et al., 2004; Benedetti et al., 1999) and muscle activation changes with brace application have been captured using electromyography (EMG) (Ramsey et al., 2007). Therefore, the relationship between medial joint contact force and knee adduction moment magnitude may not hold true with brace wear as muscle forces may change between conditions and alter the loading distribution among the soft tissues of the knee. Estimating the medial compartment contact force using an EMG-driven muscle model (Buchanan et al., 2004; Winby et al., 2009) would be a viable method for evaluating the brace function, particularly as valgus unloader bracing has been shown to reduce the muscle co-activation seen with OA (Ramsey et al., 2007). This method could also be used to verify whether or not there remains a correlation between the knee adduction moment magnitude and the medial compartment contact force when a valgus unloader brace is worn.

The No Change group did not change in knee adduction moment magnitude between conditions and had the highest magnitude of knee adduction moment of all three groups both with and without the brace. The lack of reduction in external knee adduction moment magnitude for this group may have existed because the moment was just too large for the brace to dramatically alter the kinematics of the participant. As well, this group had the highest knee rotation moments and highest range in rotation from early stance to early swing. This suggests that there may be an upper limit of the brace's ability to alter loading and rotation motion. This may be a result of the size of the brace. In contrast to Fantini Pagani et al. (2010a), the braces used in the present study are much smaller and the thigh and shank segments do not extend the entire length of the limb. Although the brace studied by Fantini Pagani et al. (2010a) may provide more support, the marketability of such a large brace would need to be evaluated.

The brace decreased the range of internal/external rotation angle over the gait cycle. The brace hinge does not allow rotation between the segments about its long axis and this effect would be exaggerated by a tight fit of the straps and thigh and shank segments around the leg. The reduction in extension moment at toe-off agrees with the results of Toriyama et al. (2011), however, there was no change in gait speed in the present study with brace application. Hence, this change is more likely a mechanical effect from brace application than a change in the braking mechanics as suggested by Toriyama et al. (2011). The reduced extension moment, along with the reduced range of internal/external rotation angle during the brace condition are different mechanical actions of the brace that potentially explain the knee joint pain relief typically seen with valgus unloader bracing. However, further study is needed to ascertain this relationship.

The brace also increased varus thrust magnitude, shown both in knee adduction angle PC3 and in the discrete measure of varus thrust. This finding was unexpected. The brace placed the participants in more abduction prior to weight acceptance and after foot-strike, but this effect did not remain after full weight bearing throughout the stance phase. Although the apparent increase in adduction angle after weight acceptance was driven by the Increase group, no group remained in increased abduction after weight acceptance. This is consistent with Otis et al. (2000) who showed no difference in frontal plane alignment during gait between a valgus unloader brace and a hinged brace with no fixed adduction between thigh and shank segments.

The maximum adduction angle during early stance for the Increase group increased during the brace condition from the no brace condition. Similarly, all participants were in more abduction prior to weight acceptance with the brace on than without. These differences were not significant ($p = 0.09$ and $p = 0.06$, respectively); however, the analysis was underpowered (power = 0.2) due to the small sample sizes of the groups. Increased adduction angle after weight acceptance was likely a primary factor in the increased knee adduction moment because of the larger moment arm from the center of pressure of the foot to the knee joint centre. An association has been shown between the presence (Chang et al., 2004) and increased magnitude of varus thrust (Kuroyanagi et al., 2011) and increased knee adduction moments. However Kuroyanagi et al. (2011) used a projection of the hip-knee-ankle angle onto the coronal plane to measure varus thrust instead of using a joint coordinate system embedded in the knee. The results would be affected by the differences in knee flexion seen between groups in this study. The increased adduction angle after weight acceptance indicates that the participants were less able to counteract the knee adduction moment with the brace on. It is uncertain whether or not the increased knee adduction moments were the cause of the increased adduction after weight acceptance, or vice versa. The cause of varus thrust is unclear, though it has been hypothesized that knee capsule and ligament deficiencies and insufficient muscle activation for knee stabilization are related to varus thrust (Chang et al., 2010). Further investigation of the passive and active stability of the knee with and without brace application should be investigated to understand the differences in adduction angle between groups. The Increase group may have decreased passive stability, possibly due to some joint space narrowing, or decreased active stability, due to an inability to develop the muscle activation patterns necessary to respond to the new abducted position at foot-strike with bracing. The other groups may be more stable due to increased muscle coordination, or because they were actively stiffening their knee through muscle co-contraction as was suggested by (Ramsey et al., 2007).

The Increase group exhibited an increased knee internal rotation moment at weight acceptance during both conditions. In recent work, Henriksen et al. (2011) has shown a positive association between the knee rotation and adduction moments. It has been hypothesized that the knee rotation moment can stress structures in the transverse plane and create shear loads that may be causative of OA (Henriksen et al., 2011). As well as increased knee adduction moments, it is possible that increased knee rotation moments

may also indicate a risk of progression, but more research needs to be done in this area. The magnitude of the knee rotation moment for the Increase group became significantly larger than the Decrease group during the brace condition, which further highlights the link between knee adduction and rotation moments. This supports the investigation of 3D kinetics to better understand the function of the valgus unloader brace. The Increase group also had increased flexion without the brace on, although the difference was small and the analysis may have been overpowered due to small variability in flexion angles between participants.

The reduced external knee adduction moment in the Decrease group was not easily explained based on the measures examined in the present study as there were no kinetic or kinematic differences between the brace and no brace condition particular to this group. However, other associative factors may be drawn out of the analysis. This group had greater flexion than the other groups for both conditions, which may indicate that these participants had higher function, however this was not supported by an increase in walking velocity or a difference in the WOMAC function scores. The Decrease group also had the smallest range in internal/external rotation between early stance and early swing for both conditions, which may signify that this group had more passive or active stiffness in the knee joint than the other groups.

The three groups had varying changes in knee adduction moment, but were not demographically different and did not have different levels of pain, function, or stiffness. However, there existed statistically significant differences among groups in the 3D mechanical response to brace application and this shows that the valgus unloader brace does not have a systematic effect on all patients. These gait adaptations should be further investigated to clarify the mechanisms by which the brace may treat OA illness and disease for different participants. As well, the gait adaptations presented in this chapter may help predict who will and will not exhibit a reduced adduction moment with brace wear.

This study did not capture individual participant satisfaction with the brace. This study will continue with a six month follow-up study that will allow for monitoring of changes in self-reported outcomes with brace wear. Measuring the differences among subgroups in self-reported outcomes would clarify the relationship between changes in knee adduction moment, pain, and function with brace wear.

A limitation of this study was that the brace mass and inertial characteristics were not

included in the inverse dynamics analysis for the brace condition. A change in the mass and moment of inertia of the limb segments between conditions may affect the kinetic results. However, given that the brace had a small mass of approximately 500g (depending on patient size) distributed over both limb segments and the inertial accelerations during stance were small, changes in kinetics were considered negligible.

In summary, there were subgroups of participants who exhibited varying changes in the knee adduction moment magnitude and other 3D kinematic and kinetic measures during gait with the immediate application of a valgus unloader brace.

There existed mechanisms alternative to reducing knee adduction moment by which the brace may effect changes in participants. Loads and motion in the transverse plane appear to be particularly important. These adaptations could help predict who will and will not benefit biomechanically from brace wear.

4.4.1 Conclusion

This study showed that the brace did not have a homogeneous effect on the knee adduction moment magnitude. The existence of groups of participants who exhibited varying changes in knee adduction moment magnitude showed that the brace may not induce a reduction in knee adduction moment magnitude for all patients. Specific 3D mechanical variables were altered with brace application in all three planes (knee flexion moment, knee rotation angle, knee adduction angle) whereas interactions occurred for four variables (knee rotation moment, knee flexion angle, knee rotation angle). The interactions indicate that non-uniform alterations in mechanics with brace application occurred among groups. These findings have specific implications for the prescription of the brace.

CHAPTER 5

CONCLUSION

5.1 Summary of Thesis Objectives

The overall aim of the thesis was to expand our understanding of the effects of a valgus unloader brace on the 3D kinetics and kinematics of the knee joint during gait. The two research objectives were:

1. To determine the immediate effect of a custom-fit valgus unloader brace on the overall magnitude of the net external knee adduction moment during gait in individuals with mild to moderate medial compartment knee OA (Chapter 3).
2. a) To identify subgroups whose overall knee adduction moment magnitude i) decreased with brace application, ii) did not change with brace application, and iii) increased with brace application (Chapter 4).
b) To determine differences in 3D knee kinematic and kinetic features after brace application and among subgroups who exhibit varying changes in net external knee adduction moment with the immediate application of a valgus unloader brace (Chapter 4).

A summary of each objective and associated results is reported below.

5.2 Summary of Chapter 3: Does the Valgus Unloader Brace Affect the External Knee Adduction Moment Magnitude?

The purpose of this chapter was to determine the immediate effect of a custom-fit valgus unloader brace on the overall magnitude of the net external knee adduction moment during gait in individuals with mild to moderate medial compartment knee OA.

The PC1 score represented the knee adduction moment magnitude and this was compared between the brace and no brace conditions. The key finding from this chapter was that there was no change in the PC1 score of the net external knee adduction moment for the overall group between the brace and no brace condition.

This indicates that the external knee adduction moment was not reduced with valgus unloader brace wear for the entire participant population. Whether a group of participants exists that decreases in knee adduction moment magnitude with brace application was the basis for Chapter 4.

5.3 Summary of Chapter 4: Does the Brace Affect All Participants in the Same Way?

The first purpose of this chapter was to identify subgroups whose overall knee adduction moment magnitude i) decreased with brace application, ii) did not change with brace application, and iii) increased with brace application. The second purpose was to determine differences in 3D knee kinematic and kinetic features after brace application and among subgroups who exhibit varying changes in net external knee adduction moment with the immediate application of a valgus unloader brace.

The key findings from this chapter were:

- There were groups of subjects who could be separated based on the change in knee adduction moment magnitude. Thirteen participants were in the Increase group and showed a statistically significant higher PC1 score for knee adduction moment (magnitude) during the brace condition than the no brace condition. Ten participants were in the Decrease group, and exhibited a statistically significant lower PC1 score for knee adduction moment (magnitude) during the brace condition the no brace condition. There was a group of 10 participants who did not change in knee

adduction moment magnitude (No Change group). The No Change group had the highest knee adduction moments of all groups in both conditions.

- There were statistically significant kinematic and kinetic effects seen with valgus unloader brace wear. There was an increase in the range of knee adduction angle at weight acceptance and between early and late stance. The brace limited external rotation of the tibia during midstance and reduced the extension moment during late stance. The brace reduced the extension moment in mid to late stance.
- There was a trend for the Increase group to have increased rotation moments at early stance and early swing, but this was not statistically significant.
- There was a statistically significant condition by group interaction for knee rotation moment magnitude. The No Change group had an increased knee rotation moment magnitude for both conditions. From the no brace to brace condition, the Increase group increased rotation moment magnitude and a statistically significantly higher knee rotation moment magnitude than the Decrease group was reported.
- There was a statistically significant condition by group interaction for knee flexion angle magnitude. The Decrease group had a higher magnitude of knee flexion angle than both groups for both conditions. The Increase group exhibited more flexion over the entire gait cycle without the brace on than with the brace on.
- There was a statistically significant condition by group interaction for knee rotation angle PC2. PC2 represented a range between rotation angles at early stance and early swing. The No Change group had a greater range than the Increase and Decrease groups for the no brace condition and a greater range than the Decrease group for the brace condition. The Increase and No Change groups had greater range during the no brace condition than the brace condition.
- There was a trend for the Increase group to exhibit a more adducted maximum angle after weight acceptance for the brace condition than the no brace condition.

The brace did not have a consistent effect on the external knee adduction moment across the participant population. It can be inferred that participants experience varying alterations to knee joint loading with brace application; however, the limitations of the

inverse dynamics analysis and its ability to provide information on the distribution of loading within the knee must be considered before making conclusions on the change to medial compartment contact force. The largest adduction and rotation loads were found in the group that did not alter the knee adduction moment with brace wear, which points at an upper limit of the brace's ability to alter loading.

Knee stability may play an important role in brace success. The Increase group was not able to counteract the increased knee adduction moments with the brace, as was evident through an increase in adduction angle after weight acceptance during the brace condition. It is uncertain whether or not the increased knee adduction moments were the cause of the increased adduction angle after weight acceptance, or vice versa. Further investigation of the passive and active stability of the knee with and without brace application should be investigated to understand the differences in maximum adduction angle between groups.

The three subgroups had varying changes in knee adduction moment, but were not demographically different and did not have different levels of pain, function, or stiffness. However, the three subgroups were different in the 3D mechanical response to brace application. This shows that the valgus unloader brace does not have a systematic effect on all patients. These gait adaptations should be further investigated to clarify the mechanisms by which the brace may treat OA illness and disease for different participants. As well, these gait adaptations may help predict who will and will not exhibit a reduced adduction moment with brace wear.

5.4 Clinical Implications

This study showed that all participants did not immediately respond with similar gait adaptations with the application of a valgus unloader brace. Although the brace has been shown to be a successful treatment of illness based on self-reported pain and function measures for the osteoarthritis population as a whole (Gaasbeek et al., 2007; Richards et al., 2005; Kirkley et al., 1999), the brace may not positively alter the biomechanical loading for all patients. There existed subgroups of participants that did not respond in the same manner to brace application as the knee adduction moment was reduced for some participants and increased for others. Patients with high knee adduction moment magnitudes may not be biomechanically affected by brace wear as the knee loads are too high for the brace to cause a significant change in loading. This may have implications for

prescription as some patients may gain from altered knee adduction moment magnitudes with brace application whereas others may not; however, participant gait adaptations at the six month lab visit need to be evaluated to understand if the changes or lack of changes in knee adduction moment magnitudes remain after an extended period of brace wear.

Participants that were in a more adducted position after weight acceptance while wearing the brace had increased knee adduction moment magnitudes. It appears that the participants in the Increase group had a less dynamically stable joint in the frontal plane. The reason for this instability is not clear from this study, but it may exist because of laxity from medial joint space narrowing, ligament laxity, or because of neuromuscular deficiencies. However, having limited motion in the frontal plane may not necessarily be an indicator of reduced knee adduction moment magnitude. Both the No Change and Decrease group did not exhibit increased adduction after weight acceptance with brace wear, but had varying changes in the knee adduction moment magnitude. The Decrease group had higher levels of flexion and were possibly higher functioning meaning that they may also have had a better ability to form new neuromuscular activation patterns when placed in a new position. It is possible that the No Change group had no change in dynamic frontal plane alignment after weight acceptance because of the high loads exhibited in two planes by this group or possibly because they were more fixed in alignment through greater active stiffening of the knee. However, this cannot be determined without further investigation of the muscle activation patterns during both conditions.

Although it is not possible as of yet to determine who may be stiffening the knee and who may exhibit better neuromuscular control, it may be possible to determine which participants exhibit increases in adduction angle with brace wear in clinic. Although varus thrust can be identified by a clinician, Kuroyanagi et al. (2011) suggested the possibility of measuring the change in knee adduction motion during weight acceptance using inexpensive digital video as a clinical tool. The presence of varus thrust was captured for several participants in this study using a digital video camera; however, the dynamic range of the thrust could not be evaluated. Markers and associated software would be needed to measure the magnitude of varus thrust in clinic.

There might be other mechanisms by which the brace reduces pain and/or slows progression of OA. The brace altered several 3D kinematic and kinetic variables of the knee joint during gait for the whole participant group and among subgroups. A reduction in

the knee extension moment and range of rotation angles was seen in the whole participant group. Among the subgroups there were differences in the loading and motion in the transverse plane, as well as differences in flexion angle magnitudes throughout the whole gait cycle. The relationship between these immediate gait adaptations and the gait adaptations measured at six months should be investigated to understand if effects of the brace remain, disappear, or change after long term brace wear and whether or not this is participant dependent. Overall 3D changes in gait mechanics need to be evaluated with respect to their effect on progression and pain so as to further develop brace design.

5.5 Limitations

There were limitations to this study. Firstly, the sample size of participants in each subgroup was low. A larger sample size would increase the power to detect significant differences and thus could clarify the differences between subgroups, particularly the difference in maximum adduction angle. Although the maximum adduction angle after weight acceptance showed a relationship with increased knee adduction moment with brace application, it is important to verify that this effect is statistically significant and repeatable.

Secondly, the brace mass and inertial characteristics were not included in the thigh and shank segment anthropometric data used in the inverse dynamics calculation during the brace condition. Changes to the mass and moment of inertia of the limb segments would affect the kinetic values of the knee. However, the brace has a mass of approximately only 500g distributed over both limb segments and the inertial accelerations during stance were small. Therefore, kinetic effects from the addition of the brace that were unaccounted using the current methods were considered negligible in this thesis. However, the change in the moment of inertia of the limb segments from the addition of a brace should be evaluated and the expected limited sensitivity of the inverse dynamics model to these changes in segment anthropometrics should be verified.

Finally, the knee adduction moment calculated using inverse dynamics needs to be interpreted with respect to joint contact loads with caution. Although the magnitude of the knee adduction moment has been associated with disease progression, the measure does not provide information about the distribution of loads among the passive and active structures of the knee. As these may alter with brace application, the limitations of using

the knee adduction moment magnitude as a surrogate measure of medial joint contact forces should be considered when interpreting the results of this thesis.

5.6 Future Research

This thesis provided valuable information concerning the 3D effects of valgus unloader brace application during gait on medial compartment knee OA participants. Data suggesting that not all OA patients respond with the same gait adaptations to brace application and that there exist other 3D dynamic changes in the knee joint mechanics with brace application were presented. This is an early step in understanding the 3D effects of valgus unloader brace function and brings about various research questions that could be addressed with future studies. Specifically, research should be focused in three areas: understanding the relationships between 3D kinematic and kinetic gait variables and biomechanical OA progression, understanding the mechanisms by which the brace relieves the symptoms of OA illness like pain and function, and further understanding the mechanisms by which the brace may affect OA disease progression with particular focus on the utilization of muscle modeling.

- In the study of knee OA, the knee adduction moment calculated with inverse dynamics provides a limited estimation of the medial compartment contact force as it gives no indication of increased compartment forces due to co-activation of muscles that exists within the OA population (Hubley-Kozey et al., 2008; Childs et al., 2004; Benedetti et al., 1999). Therefore muscle activation changes captured with electromyography (EMG) that occur with brace application should be evaluated to gain a better understanding on knee loading. Estimating the medial compartment contact force using an EMG-driven muscle model (Buchanan et al., 2004; Winby et al., 2009) would be a strong method for calculating changes in contact loads that occur with brace application.
- It is also important to gain an understanding of what levels of medial compartment forces will increase the disease progression of OA and whether or not the brace can provide that reduction. Capturing an estimation of knee joint compartmental forces using EMG-driven muscle modeling would provide strong estimates of the force in the medial compartment of the knee. Deriving the relationship between

progression and medial compartment contact forces through a longitudinal study of OA progression would capture whether there is a threshold to define unhealthy loads and whether or not the brace can reduce the load below that threshold.

- There is currently little available literature regarding rotation moment and angles in the illness and disease process of OA. Investigation should be done into the role these variables may play in biomechanical progression and joint pain relief. If these prove to be important factors, whether or not the brace can improve patient symptoms and progression via altered rotational loading and motion should be investigated.
- Stability between participant subgroups should be addressed. Relating differences among subgroups in the active stability (neuromuscular) and passive stability (capsulo-ligamentous (Chang et al., 2010)) to adduction angle measures could identify if some participants are at greater risk of increased adducted alignment with brace wear or if some participants could be trained to adapt to brace wear with neuromuscular coordination.
- Braces have been shown to treat the illness of OA; however, understanding the relationship between successful treatment of illness and successful treatment of disease would more thoroughly explain brace function. The relationship between self-reported pain and function outcomes and OA disease progression with valgus unloader brace wear would clarify if there exist patients who benefit from pain relief, without benefiting from slowing of progression. A longitudinal study of disease progression in participants with a valgus unloader brace would provide this information.
- As the three subgroups of participants exhibited kinematic and kinetic differences, it may be possible to develop a set of specific gait characteristics that would predict which participants will show a reduction in the knee adduction moment with valgus unloader brace application. A larger sample size may identify other differences in gait mechanics or demographic information among subgroups. This could have implications on the use of brace wear to slow biomechanical progression.

5.7 Conclusion

This thesis expanded the current understanding of the effects of a valgus unloader brace on dynamic 3D kinetic and kinematic measures of knee joint function during gait. It showed that not all OA participants responded with the same gait adaptations to brace application and that there existed other 3D dynamic changes in the knee joint mechanics. This was also the first study that related changes in 3D kinematic and kinetic variables to changes in the knee adduction moment magnitude with brace application.

It was found that valgus unloader brace wear did not affect the knee adduction moment magnitude of the participant group in a consistent manner. Participants with high knee adduction moments were not significantly affected by brace wear which shows that the brace may have an upper limit of its ability to reduce loads. There were identifiable kinematic differences between groups who showed a positive and negative biomechanical response to valgus unloader brace wear. An increase in abduction angle after weight acceptance was related to an increased knee adduction moment. A decrease in knee adduction moment was related to increased flexion over the entire gait cycle and a smaller range in internal/external rotation with and without the brace on. Loads and motion in the transverse plane should be further investigated to understand their importance in illness and disease progression of knee OA and the effect of the brace on knee OA. The kinematic and kinetic differences between groups of participants may have implications on prescription and for the development of patient specific treatments.

Future research should be done to understand the mechanisms by which the brace relieves the symptoms of OA illness, such as pain and function and to further understand the mechanisms by which the brace may affect OA disease progression.

APPENDIX A

LOADING VECTORS

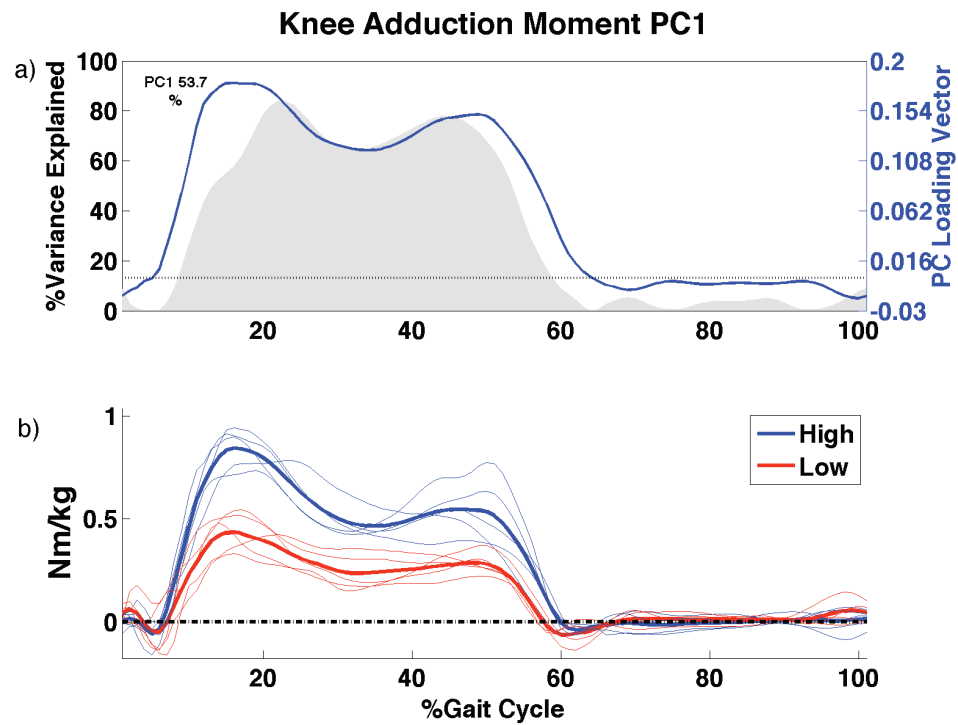


Figure A.1: Knee adduction moment PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores are shown in (b). Blue lines are KAM waveforms with high scores, red lines are KAM waveforms with low scores, thin lines are the top or bottom fifth percentile of KAM waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms ((+) is adduction, (-) is abduction).

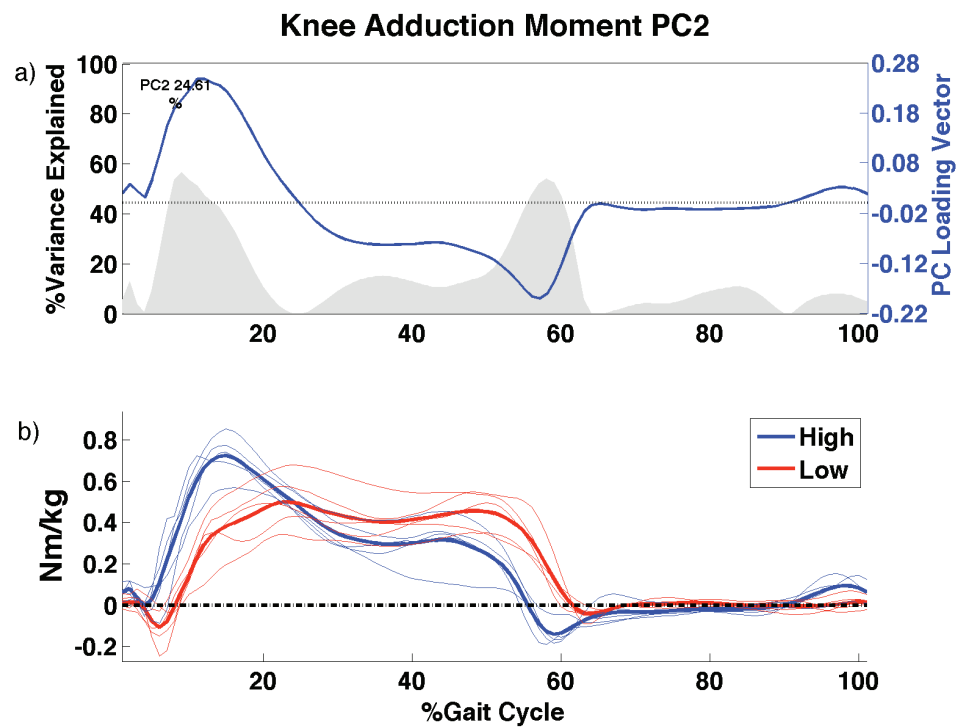


Figure A.2: Knee adduction moment PC2. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC2-scores are shown in (b). Blue lines are KAM waveforms with high scores, red lines are KAM waveforms with low scores, thin lines are the top or bottom fifth percentile of KAM waveforms based on their PC2-score, and thick lines are the average of the five high and the five low PC2 waveforms ((+) is adduction, (-) is abduction).

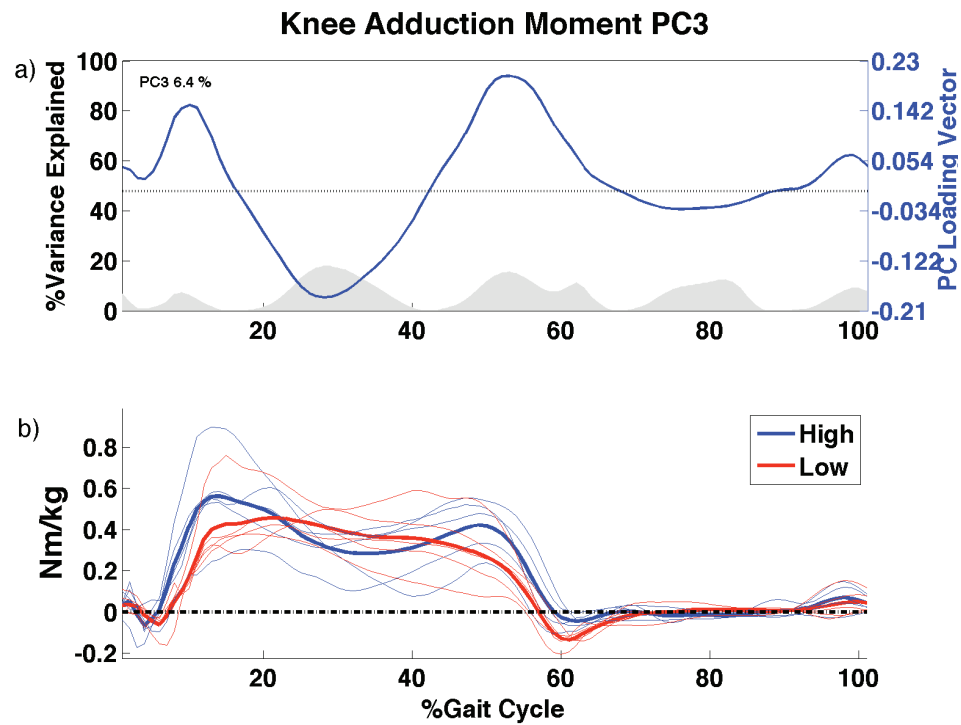


Figure A.3: Knee adduction moment PC3. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC3-scores are shown in (b). Blue lines are KAM waveforms with high scores, red lines are KAM waveforms with low scores, thin lines are the top or bottom fifth percentile of KAM waveforms based on their PC3-score, and thick lines are the average of the five high and the five low PC3 waveforms ((+) is adduction, (-) is abduction).

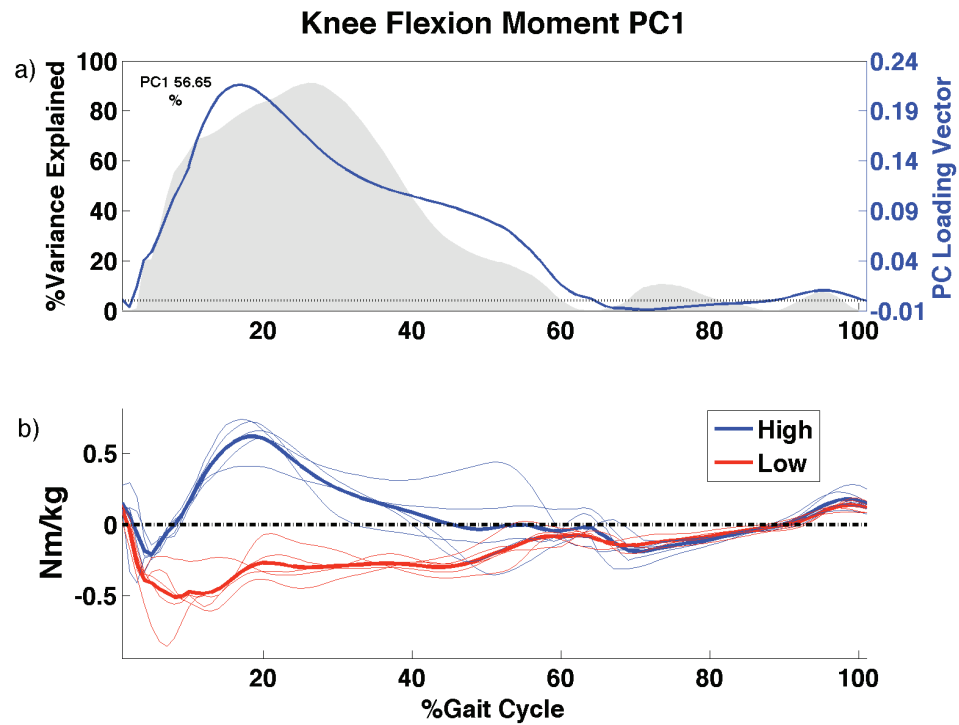


Figure A.4: Knee flexion moment PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores are shown in (b). Blue lines are KFM waveforms with high scores, red lines are KFM waveforms with low scores, thin lines are the top or bottom fifth percentile of KFM waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms ((+) flexion, (-) extension).

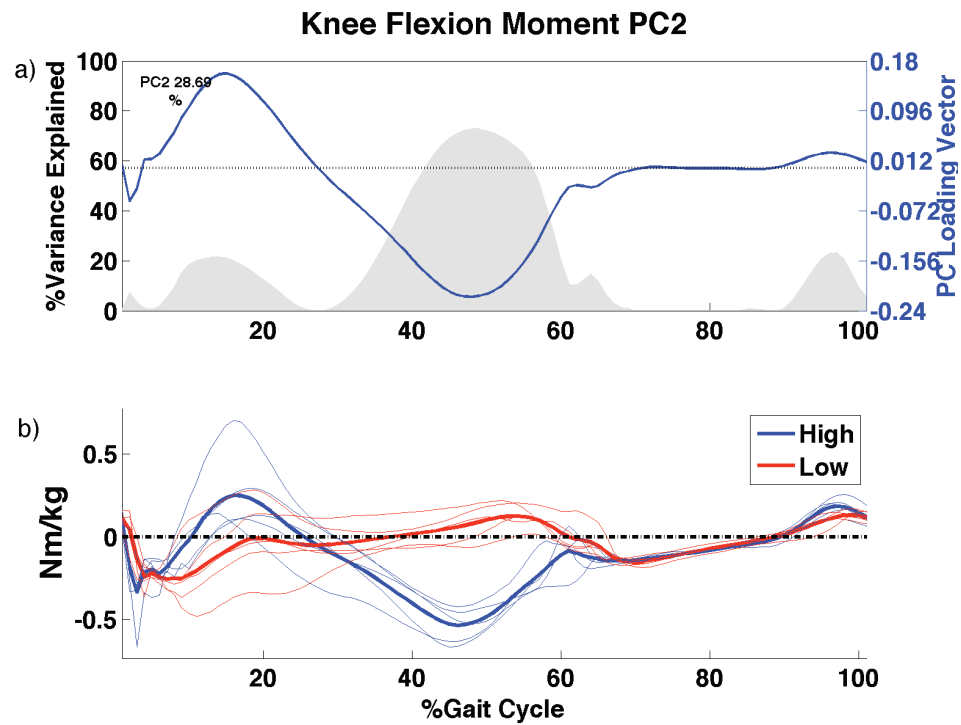


Figure A.5: Knee flexion moment PC2. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC2-scores are shown in (b). Blue lines are KFM waveforms with high scores, red lines are KFM waveforms with low scores, thin lines are the top or bottom fifth percentile of KFM waveforms based on their PC2-score, and thick lines are the average of the five high and the five low PC2 waveforms ((+) flexion, (-) extension).

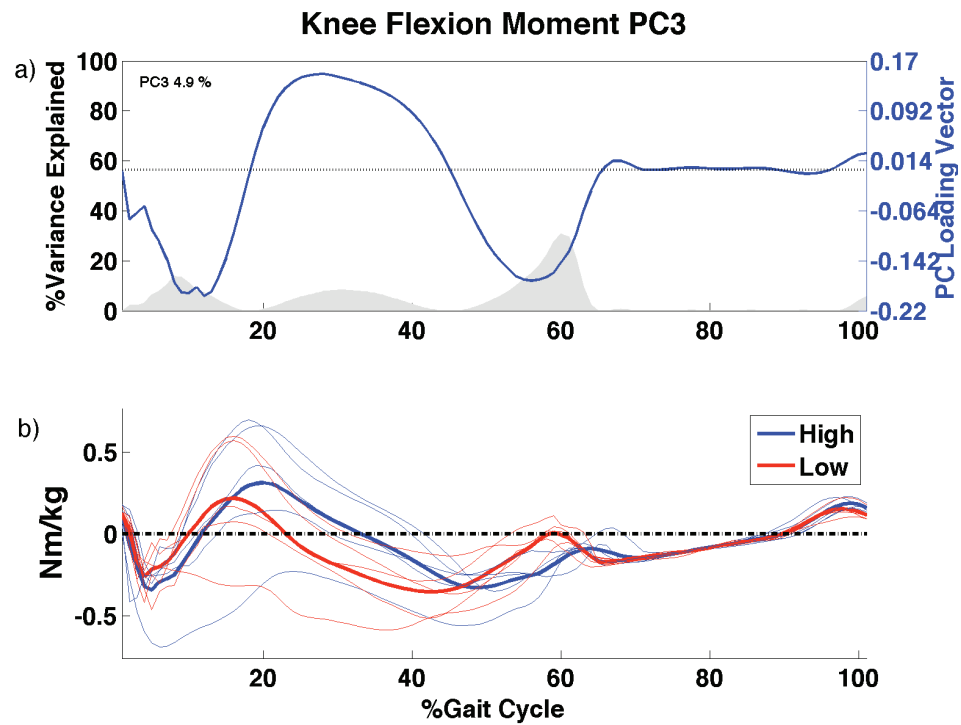


Figure A.6: Knee flexion moment PC3. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC3-scores are shown in (b). Blue lines are KFM waveforms with high scores, red lines are KFM waveforms with low scores, thin lines are the top or bottom fifth percentile of KFM waveforms based on their PC3-score, and thick lines are the average of the five high and the five low PC3 waveforms ((+) flexion, (-) extension).

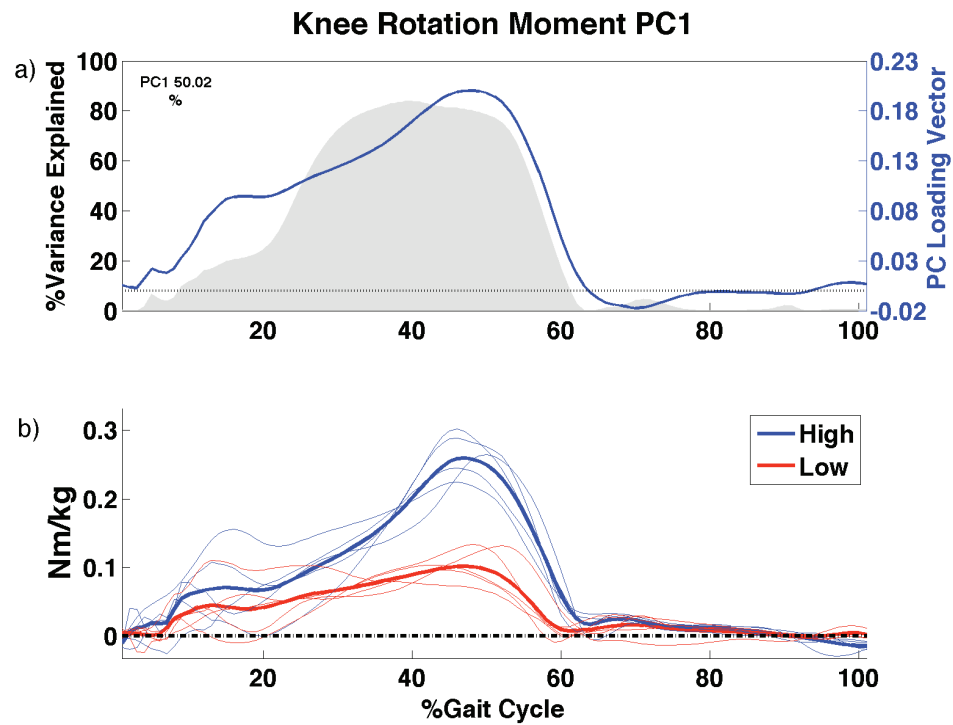


Figure A.7: Knee rotation moment PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores are shown in (b). Blue lines are KRM waveforms with high scores, red lines are KRM waveforms with low scores, thin lines are the top or bottom fifth percentile of KRM waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms ((+) internal rotation acting on tibia, (-) external rotation moment acting on tibia).

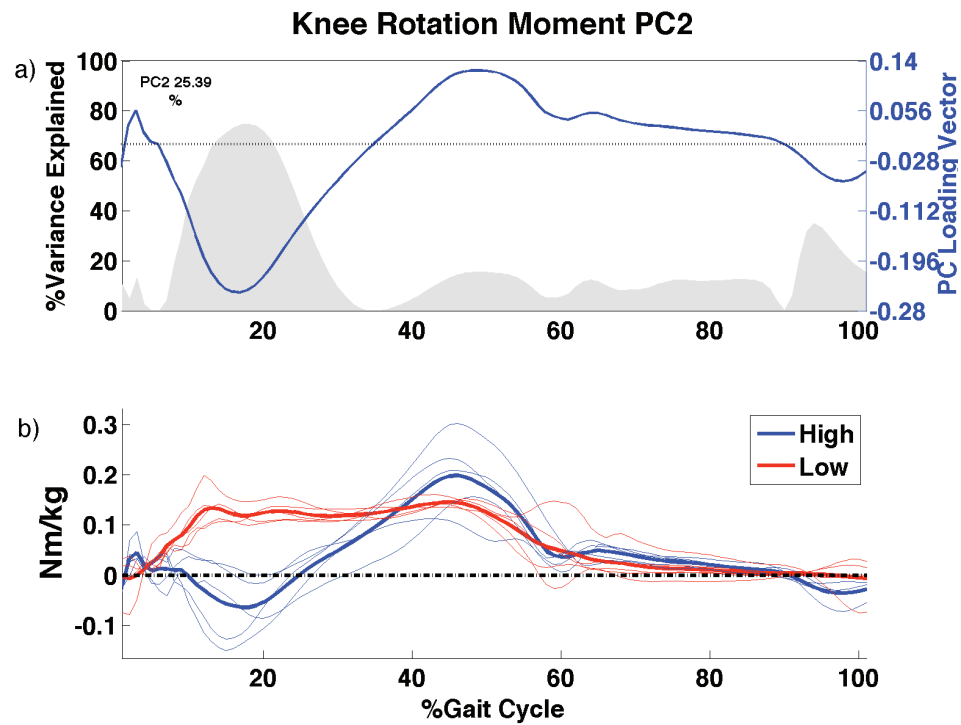


Figure A.8: Knee rotation moment PC2. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC2-scores are shown in (b). Blue lines are KRM waveforms with high scores, red lines are KRM waveforms with low scores, thin lines are the top or bottom fifth percentile of KRM waveforms based on their PC2-score, and thick lines are the average of the five high and the five low PC2 waveforms ((+) internal rotation acting on tibia, (-) external rotation moment acting on tibia).

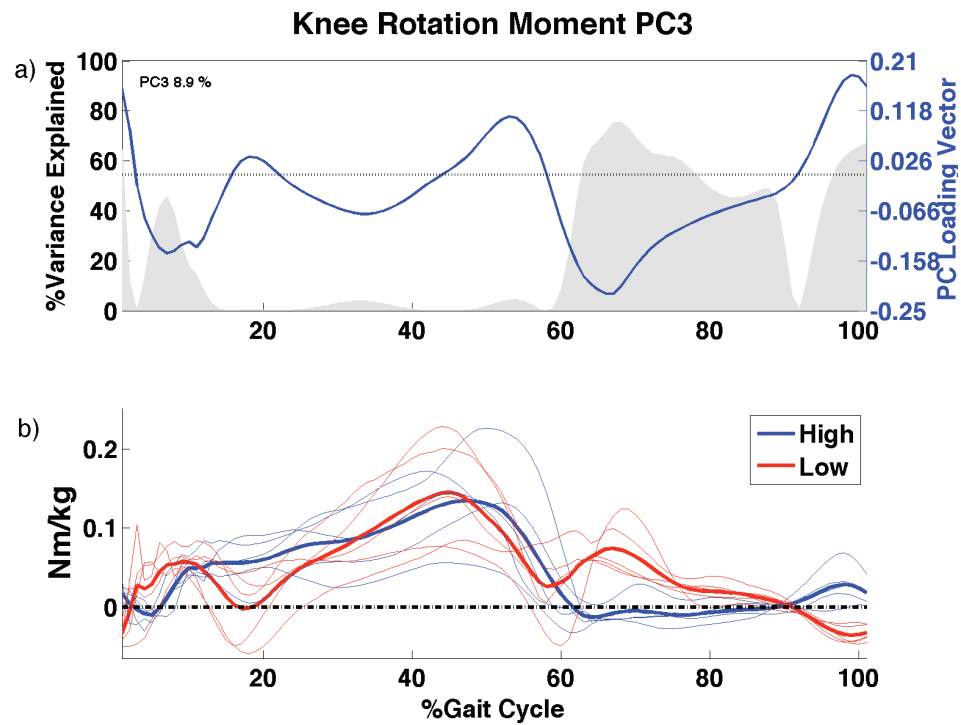


Figure A.9: Knee rotation moment PC3. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC3-scores are shown in (b). Blue lines are KRM waveforms with high scores, red lines are KRM waveforms with low scores, thin lines are the top or bottom fifth percentile of KRM waveforms based on their PC3-score, and thick lines are the average of the five high and the five low PC3 waveforms ((+) internal rotation acting on tibia, (-) external rotation moment acting on tibia).

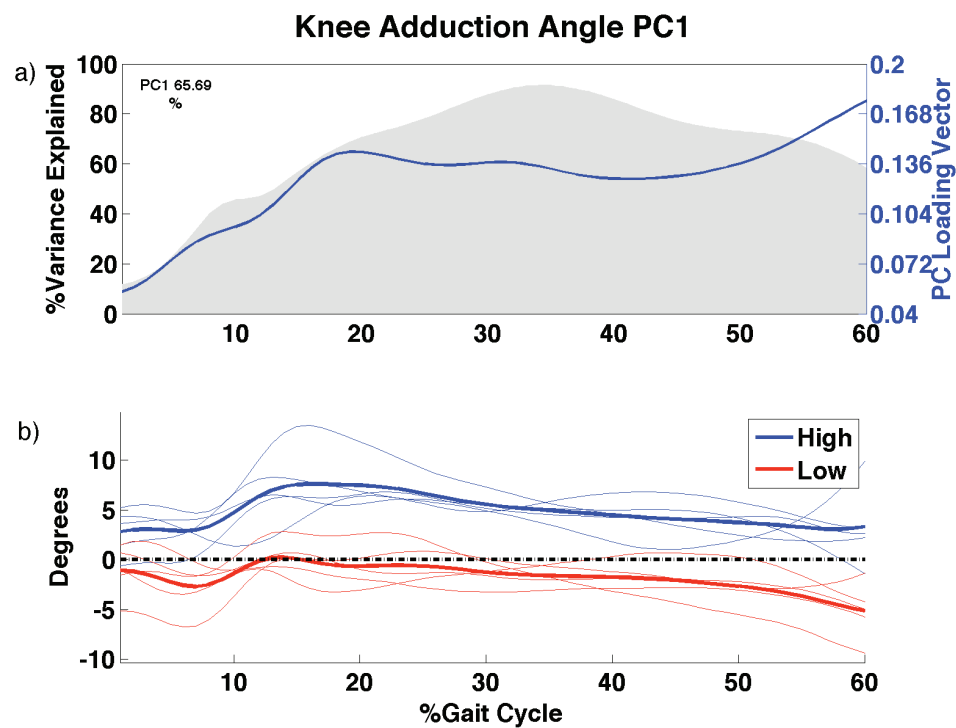


Figure A.10: Knee adduction angle PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores are shown in (b). Blue lines are KAA waveforms with high scores, red lines are KAA waveforms with low scores, thin lines are the top or bottom fifth percentile of KAA waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms ((+) is adduction, (-) is abduction).

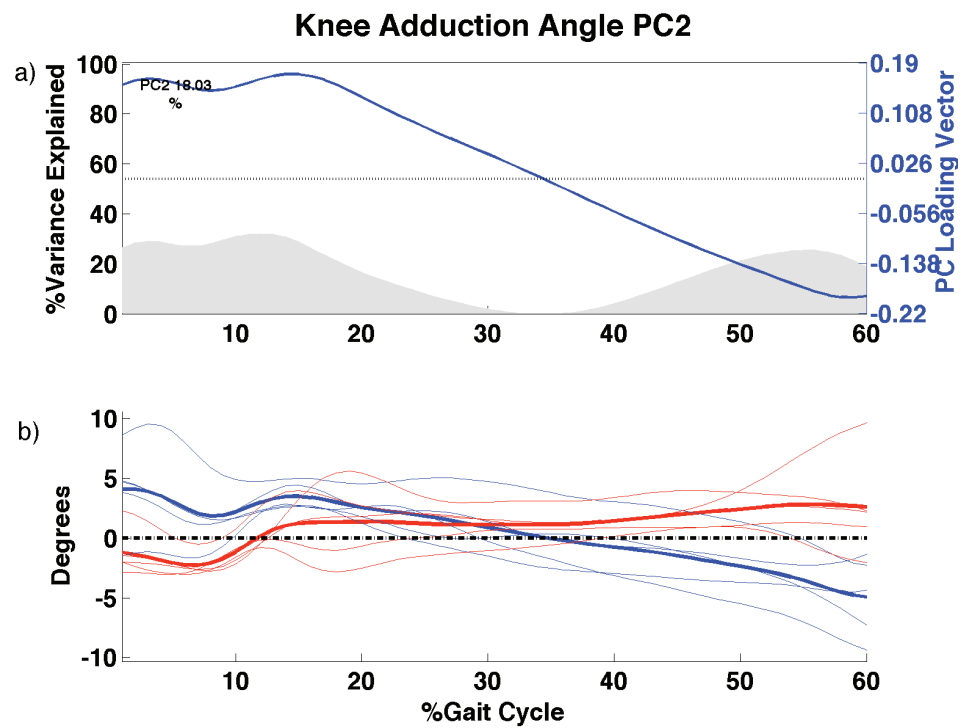


Figure A.11: Knee adduction angle PC2. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC2-scores are shown in (b). Blue lines are KAA waveforms with high scores, red lines are KAA waveforms with low scores, thin lines are the top or bottom fifth percentile of KAA waveforms based on their PC2-score, and thick lines are the average of the five high and the five low PC2 waveforms ((+) is adduction, (-) is abduction).

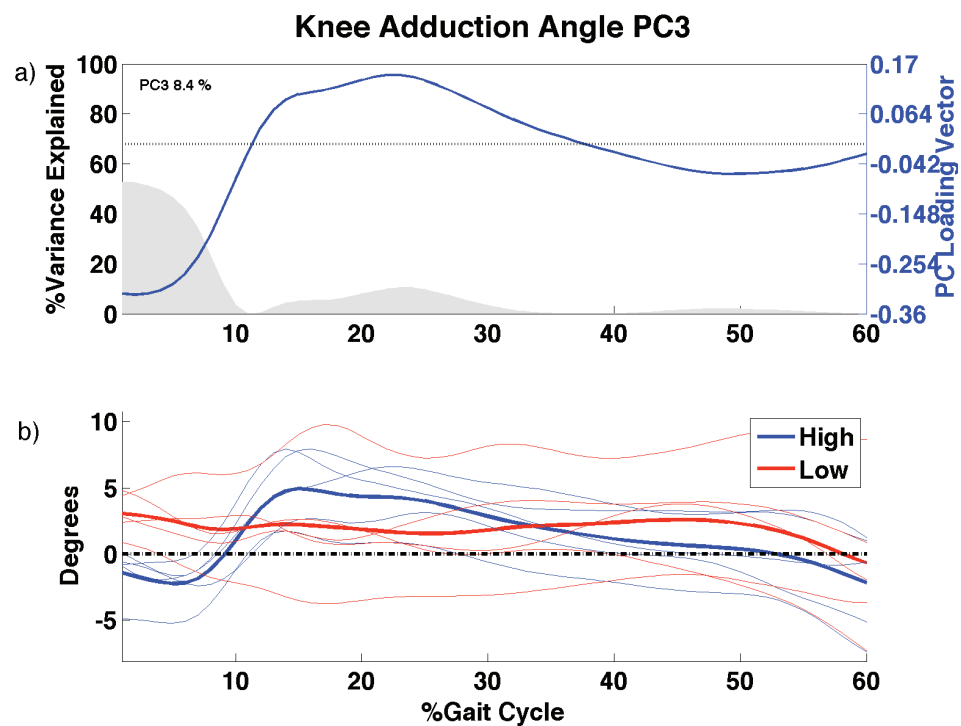


Figure A.12: Knee adduction angle PC3. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC3-scores are shown in (b). Blue lines are KAA waveforms with high scores, red lines are KAA waveforms with low scores, thin lines are the top or bottom fifth percentile of KAA waveforms based on their PC3-score, and thick lines are the average of the five high and the five low PC3 waveforms ((+) is adduction, (-) is abduction).

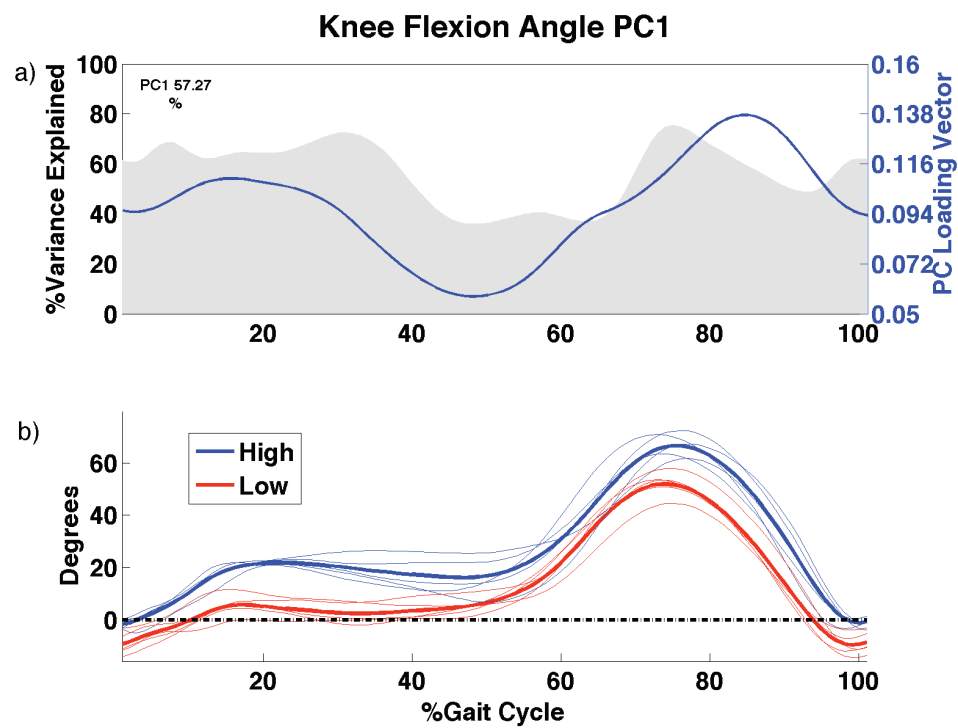


Figure A.13: Knee flexion angle PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores are shown in (b). Blue lines are KFA waveforms with high scores, red lines are KFA waveforms with low scores, thin lines are the top or bottom fifth percentile of KFA waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms ((+) flexion, (-) extension).

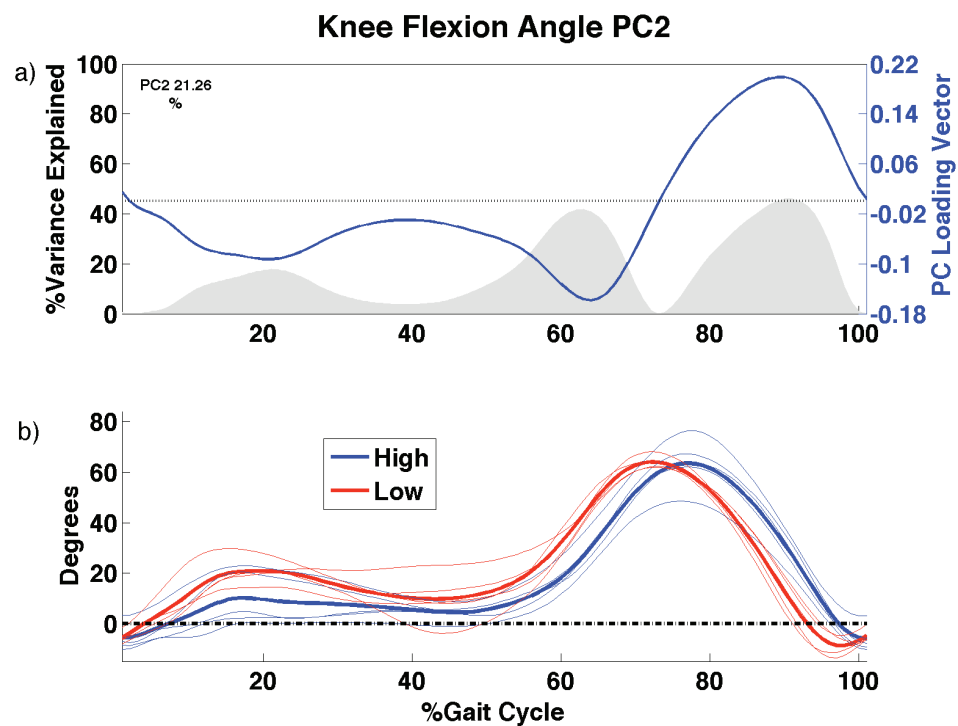


Figure A.14: Knee flexion angle PC2. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC2-scores are shown in (b). Blue lines are KFA waveforms with high scores, red lines are KFA waveforms with low scores, thin lines are the top or bottom fifth percentile of KFA waveforms based on their PC2-score, and thick lines are the average of the five high and the five low PC2 waveforms ((+) flexion, (-) extension).

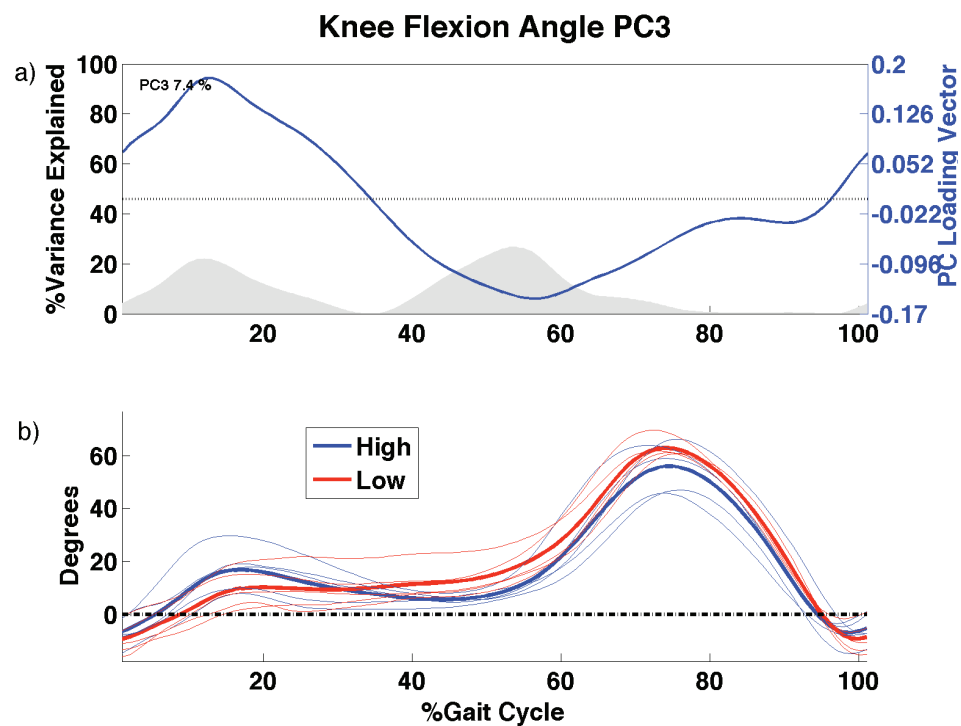


Figure A.15: Knee flexion angle PC3. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC3-scores are shown in (b). Blue lines are KFA waveforms with high scores, red lines are KFA waveforms with low scores, thin lines are the top or bottom fifth percentile of KFA waveforms based on their PC3-score, and thick lines are the average of the five high and the five low PC3 waveforms ((+) flexion, (-) extension).

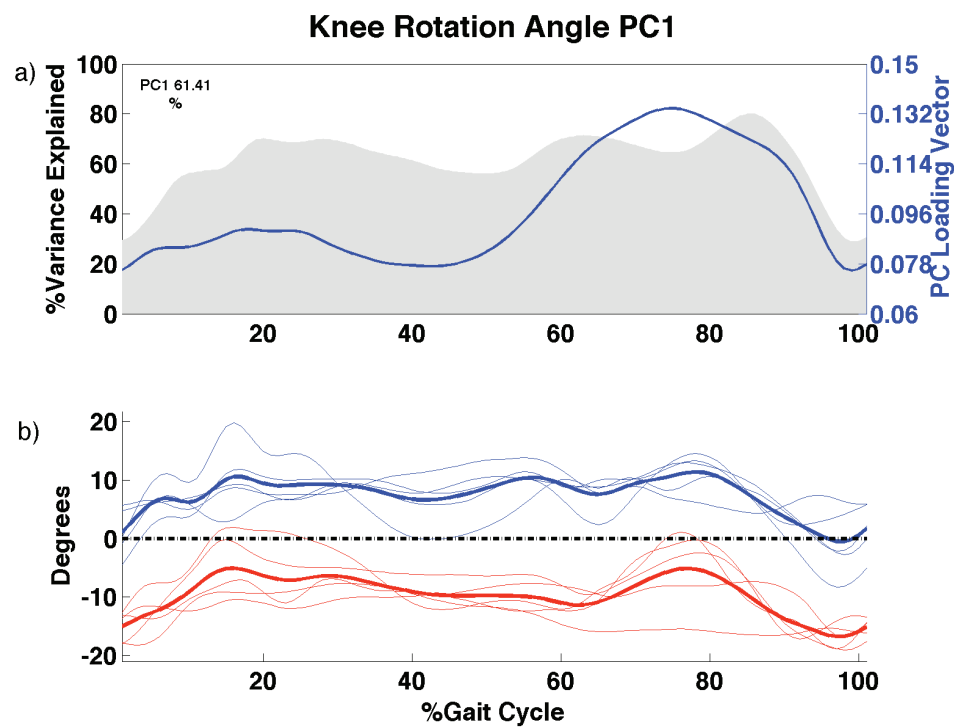


Figure A.16: Knee rotation angle PC1. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC1-scores are shown in (b). Blue lines are KRA waveforms with high scores, red lines are KRA waveforms with low scores, thin lines are the top or bottom fifth percentile of KRA waveforms based on their PC1-score, and thick lines are the average of the five high and the five low PC1 waveforms ((+) internal rotation of tibia, (-) external rotation of tibia).

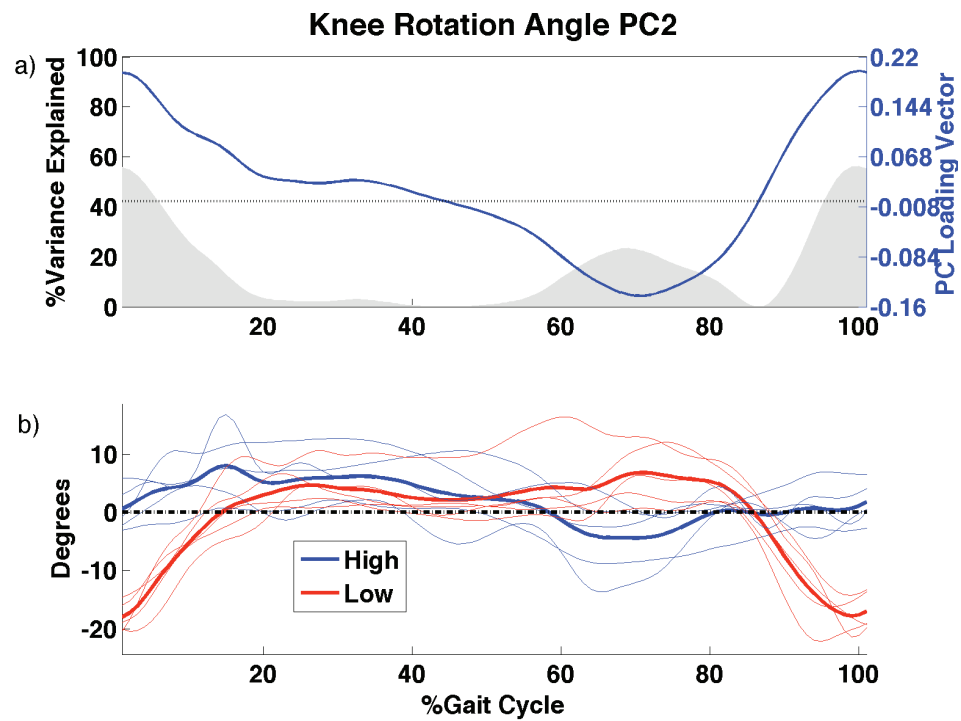


Figure A.17: Knee rotation angle PC2. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC2-scores are shown in (b). Blue lines are KRA waveforms with high scores, red lines are KRA waveforms with low scores, thin lines are the top or bottom fifth percentile of KRA waveforms based on their PC2-score, and thick lines are the average of the five high and the five low PC2 waveforms ((+) internal rotation of tibia, (-) external rotation of tibia).

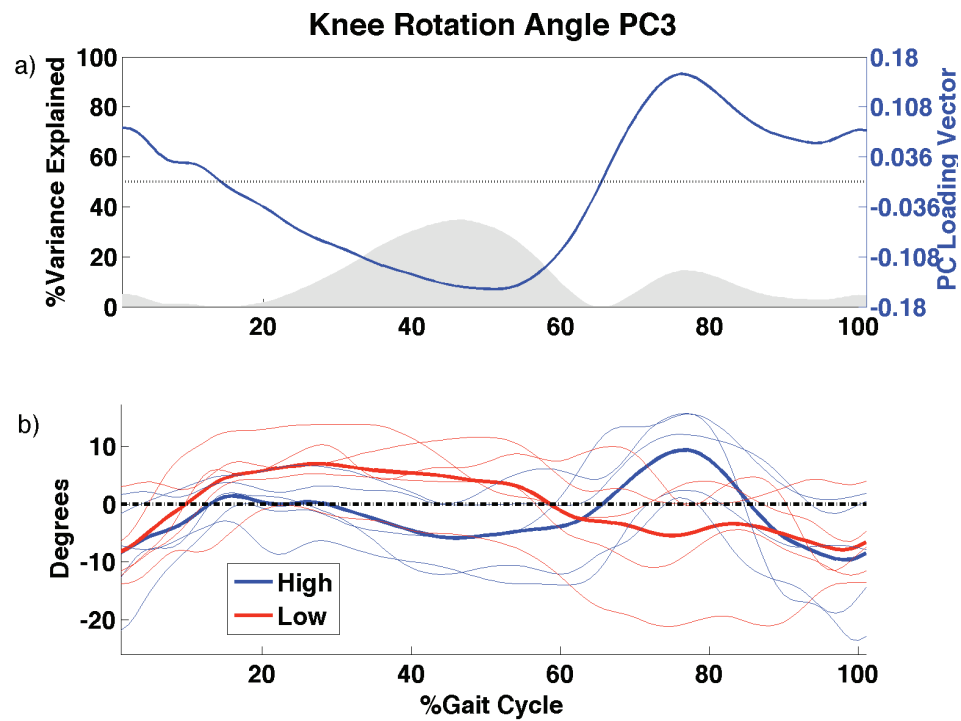


Figure A.18: Knee rotation angle PC3. The loading vector is shown in (a). The percentage of the scaled variance explained is shown in the shaded area beneath the loading vector curve (a). Five waveforms each with high and low PC3-scores are shown in (b). Blue lines are KRA waveforms with high scores, red lines are KRA waveforms with low scores, thin lines are the top or bottom fifth percentile of KRA waveforms based on their PC3-score, and thick lines are the average of the five high and the five low PC3 waveforms ((+) internal rotation of tibia, (-) external rotation of tibia).

APPENDIX B

VARUS THRUST WAVEFORMS OF PARTICIPANTS REMOVED FROM THE ANALYSIS

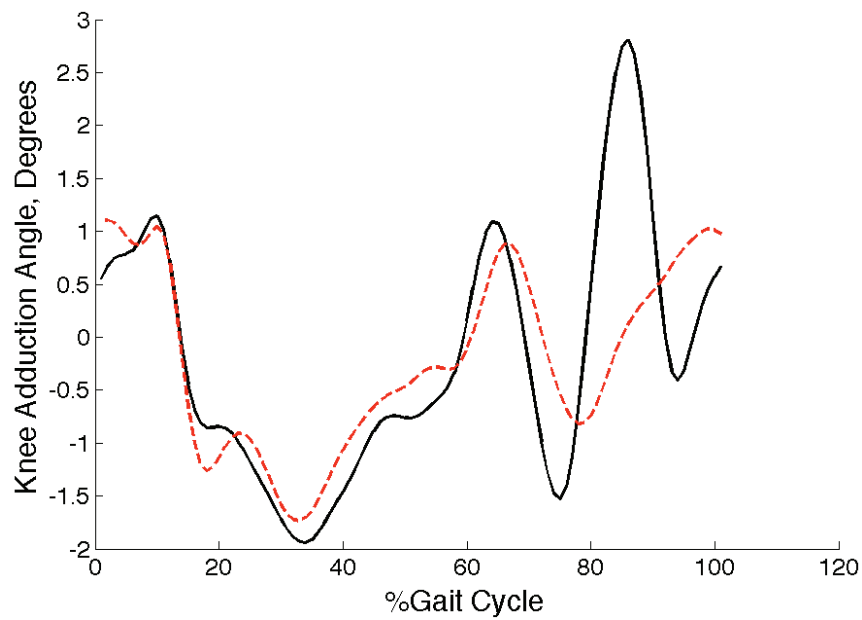


Figure B.1: Knee adduction angle waveform of participant in the Decrease group removed from varus thrust analysis. This participant did not exhibit varus thrust according to the thresholds set in this thesis.

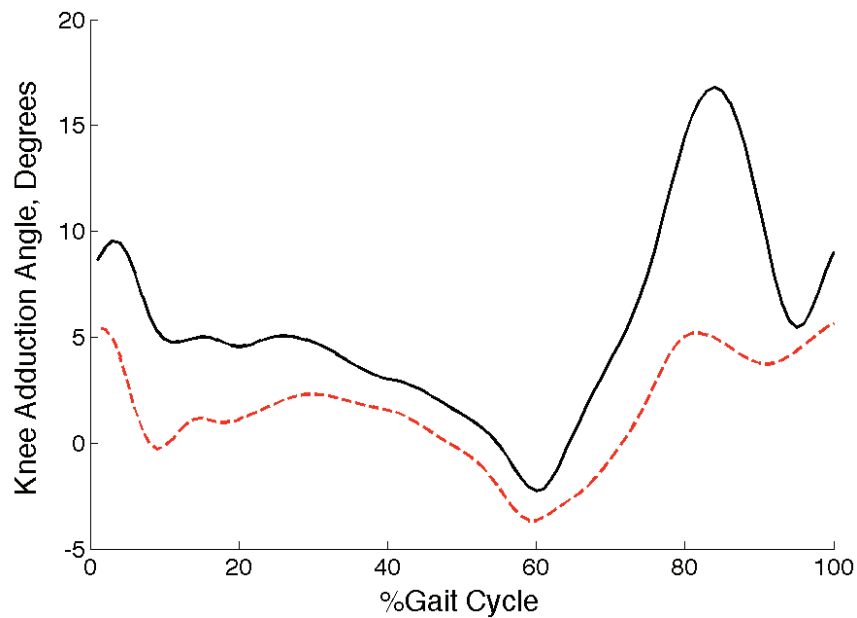


Figure B.2: Knee adduction angle waveform of participant in the Increase group removed from varus thrust analysis. This participant did not exhibit varus thrust according to the thresholds set in this thesis.

BIBLIOGRAPHY

- Altman, R., Asch, E., Bloch, D., Bole, G., Borenstein, D., Brandt, K., Christy, W., Cooke, T., Greenwald, R., Hochberg, M., et al. (1986). Development of criteria for the classification and reporting of osteoarthritis: classification of osteoarthritis of the knee. *Arthritis & Rheumatism*, 29(8):1039–1049.
- Andriacchi, T. and Mündermann, A. (2006). The role of ambulatory mechanics in the initiation and progression of knee osteoarthritis. *Current Opinion in Rheumatology*, 18(5):514–518.
- Andriacchi, T., Mündermann, A., Smith, R., Alexander, E., Dyrby, C., and Koo, S. (2004). A framework for the in vivo pathomechanics of osteoarthritis at the knee. *Annals of Biomedical Engineering*, 32(3):447–457.
- Astephen, J. and Deluzio, K. (2005). Changes in frontal plane dynamics and the loading response phase of the gait cycle are characteristic of severe knee osteoarthritis: application of a multidimensional analysis technique. *Clinical Biomechanics*, 20(2):209–217.
- Astephen, J., Deluzio, K., Caldwell, G., and Dunbar, M. (2008). Biomechanical changes at the hip, knee, and ankle joints during gait are associated with knee osteoarthritis severity. *Journal of Orthopaedic Research*, 26(3):332–341.
- Astephen Wilson, J., Deluzio, K., Dunbar, M., Caldwell, G., and Hubley-Kozey, C. (2011). The association between knee joint biomechanics and neuromuscular control and moderate knee osteoarthritis radiographic and pain severity. *Osteoarthritis and Cartilage*, 19(2):186–193.
- Baliunas, A., Hurwitz, D., Ryals, A., Karrar, A., Case, J., Block, J., and Andriacchi, T. (2002). Increased knee joint loads during walking are present in subjects with knee osteoarthritis. *Osteoarthritis and Cartilage*, 10(7):573–579.
- Bellamy, N., Buchanan, W., Goldsmith, C., Campbell, J., and Stitt, L. (1988). Validation study of WOMAC: a health status instrument for measuring clinically important patient relevant outcomes to antirheumatic drug therapy in patients with osteoarthritis of the hip or knee. *The Journal of Rheumatology*, 15(12):1833–1840.
- Benedetti, M., Bonato, P., Catani, F., D’Alessio, T., Knaflitz, M., Marcacci, M., and Simoncini, L. (1999). Myoelectric activation pattern during gait in total knee replacement: relationship with kinematics, kinetics, and clinical outcome. *IEEE Transactions on Rehabilitation Engineering*, 7(2):140–149.
- Biden, E., O’Connor, J., and Collins, J. (1990). Gait analysis. In Daniel, D., editor, *Knee Ligaments: Structure, Function, Injury and Repair*, pages 291–293. Raven Press Ltd.

- Brandt, K., Radin, E., Dieppe, P., and Van De Putte, L. (2006). Yet more evidence that osteoarthritis is not a cartilage disease. *Annals of the Rheumatic Diseases*, 65(10):1261–1264.
- Buchanan, T., Lloyd, D., Manal, K., and Besier, T. (2004). Neuromusculoskeletal modeling: estimation of muscle forces and joint moments and movements from measurements of neural command. *Journal of Applied Biomechanics*, 20(4):367–395.
- Buckwalter, J. and Lappin, D. (2000). The disproportionate impact of chronic arthralgia and arthritis among women. *Clinical Orthopaedics and Related Research*, 372:159–168.
- Buckwalter, J., Stanish, W., Rosier, R., Schenck Jr, R., Dennis, D., and Coutts, R. (2001). The increasing need for nonoperative treatment of patients with osteoarthritis. *Clinical Orthopaedics and Related Research*, 385:36–45.
- Canadian Institute for Health Information (2009). *Canadian Institute for Health Information, Hip and Knee Replacements in Canada – Canadian Joint Replacement Registry (CJRR) 2008-2009 Annual Report*. CIHI, Ottawa, Ontario.
- Canadian Institute for Health Information (2010). Canadian Joint Replacement Registry. <http://www.cihi.ca/cjrr/>.
- Chang, A., Hayes, K., Dunlop, D., Hurwitz, D., Song, J., Cahue, S., Genge, R., and Sharma, L. (2004). Thrust during ambulation and the progression of knee osteoarthritis. *Arthritis & Rheumatism*, 50(12):3897–3903.
- Chang, A., Hochberg, M., Song, J., Dunlop, D., Chmiel, J., Nevitt, M., Hayes, K., Eaton, C., Bathon, J., Jackson, R., et al. (2010). Frequency of varus and valgus thrust and factors associated with thrust presence in persons with or at higher risk of developing knee osteoarthritis. *Arthritis & Rheumatism*, 62(5):1403–1411.
- Childs, J., Sparto, P., Fitzgerald, G., Bizzini, M., and Irrgang, J. (2004). Alterations in lower extremity movement and muscle activation patterns in individuals with knee osteoarthritis. *Clinical Biomechanics*, 19(1):44–49.
- Costigan, P., Wyss, U., Deluzio, K., and Li, J. (1992). Semiautomatic three-dimensional knee motion assessment system. *Medical and Biological Engineering and Computing*, 30(3):343–350.
- Cresswell, A., Oddsson, L., and Thorstensson, A. (1994). The influence of sudden perturbations on trunk muscle activity and intra-abdominal pressure while standing. *Experimental Brain Research*, 98(2):336–341.
- Crowninshield, R. and Brand, R. (1981). The prediction of forces in joint structures: distribution of intersegmental resultants. *Exercise and Sport Sciences Reviews*, 9(1):159–181.
- Davidson, P. L., Sanderson, D. J., and Loomer, R. L. (1998). Kinematics of valgus bracing for medial gonarthrosis: technical report. *Clinical Biomechanics*, 13(6):414–419.

- De Vita, P. and Hortobagyi, T. (2001). Functional knee brace alters predicted knee muscle and joint forces in people with ACL reconstruction during walking. *Journal of Applied Biomechanics*, 17(4):297–311.
- Deluzio, K. and Astephen, J. (2007). Biomechanical features of gait waveform data associated with knee osteoarthritis. An application of principal component analysis. *Gait & Posture*, 25(1):86–93.
- Dunbar, M., Howard, A., Bogoch, E., Parvizi, J., and Kreder, H. (2009). Orthopaedics in 2020: predictors of musculoskeletal need. *The Journal of Bone and Joint Surgery. American volume*, 91(9):2276–2286.
- Erhart, J., Dyrby, C., D’Lima, D., Colwell, C., and Andriacchi, T. (2010). Changes in in vivo knee loading with a variable-stiffness intervention shoe correlate with changes in the knee adduction moment. *Journal of Orthopaedic Research*, 28(12):1548–1553.
- Fantini Pagani, C., Bohle, C., Potthast, W., and Brüggemann, G. (2010a). Short-term effects of a dedicated knee orthosis on knee adduction moment, pain, and function in patients with osteoarthritis. *Archives of Physical Medicine and Rehabilitation*, 91(12):1936–1941.
- Fantini Pagani, C., Potthast, W., and Brüggemann, G. (2010b). The effect of valgus bracing on the knee adduction moment during gait and running in male subjects with varus alignment. *Clinical Biomechanics*, 25(1):70–76.
- Felson, D., Naimark, A., Anderson, J., Kazis, L., Castelli, W., and Meenan, R. (1987). The prevalence of knee osteoarthritis in the elderly. The Framingham osteoarthritis study. *Arthritis & Rheumatism*, 30(8):914–918.
- Felson, D. and Radin, E. (1994). What causes knee osteoarthritis: are different compartments susceptible to different risk factors? *Journal of Rheumatology*, 21(2):181–183.
- Gaasbeek, R., Groen, B., Hampsink, B., van Heerwaarden, R., and Duysens, J. (2007). Valgus bracing in patients with medial compartment osteoarthritis of the knee: A gait analysis study of a new brace. *Gait & Posture*, 26(1):3–10.
- Gok, H., Ergin, S., and Yavuzer, G. (2002). Kinetic and kinematic characteristics of gait in patients with medial knee arthrosis. *Acta Orthopaedica Scandinavica*, 73(6):647–652.
- Grood, E. and Suntay, W. (1983). A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *Journal of Biomechanical Engineering*, 105:136–144.
- Hannan, M., Felson, D., and Pincus, T. (2000). Analysis of the discordance between radiographic changes and knee pain in osteoarthritis of the knee. *The Journal of Rheumatology*, 27(6):1513–1517.

- Hatfield, G., Hubley-Kozey, C., Astephen Wilson, J., and Dunbar, M. (2010). The effect of total knee arthroplasty on knee joint kinematics and kinetics during gait. *The Journal of Arthroplasty*, 26(2):309–318.
- Hawker, G., French, M., Waugh, E., Gignac, M., Cheung, C., and Murray, B. (2010). The multidimensionality of sleep quality and its relationship to fatigue in older adults with painful osteoarthritis. *Osteoarthritis and Cartilage*, 18:1365–1371.
- Hawker, G., Gignac, M., Badley, E., Davis, A., French, M., Li, Y., Perruccio, A., Power, J., Sale, J., and Lou, W. (2011). A longitudinal study to explain the pain-depression link in older adults with osteoarthritis. *Arthritis Care & Research*, 63(10):1382–1390.
- Health Canada (2003). *Arthritis in Canada. An ongoing challenge*. Health Canada, Ottawa, Ontario.
- Henriksen, M., Bennell, K., Hinman, R., Metcalf, B., Ann-Bowles, K., and Creaby, M. (2011). The knee rotation moment. A new pathomechanical paradigm in OA? In *Abstracts of the 2011 World Congress on Osteoarthritis*, volume 19S1, pages S89–S90. Osteoarthritis and Cartilage.
- Hewett, T., Noyes, F., Barber-Westin, S., and Heckmann, T. (1998). Decrease in knee joint pain and increase in function in patients with medial compartment arthrosis: a prospective analysis of valgus bracing. *Orthopedics*, 21(2):131–140.
- Hubley-Kozey, C., Deluzio, K., and Dunbar, M. (2008). Muscle co-activation patterns during walking in those with severe knee osteoarthritis. *Clinical Biomechanics*, 23(1):71–80.
- Hubley-Kozey, C., Deluzio, K., Landry, S., McNutt, J., and Stanish, W. (2006). Neuromuscular alterations during walking in persons with moderate knee osteoarthritis. *Journal of Electromyography and Kinesiology*, 16(4):365–378.
- Hurwitz, D., Ryals, A., Case, J., Block, J., and Andriacchi, T. (2002). The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. *Journal of Orthopaedic Research*, 20(1):101–107.
- Jain, N., Higgins, L., Ozumba, D., Guller, U., Cronin, M., Pietrobon, R., and Katz, J. (2005). Trends in epidemiology of knee arthroplasty in the United States, 1990–2000. *Arthritis & Rheumatism*, 52(12):3928–3933.
- Jordan, J., Helmick, C., Renner, J., Luta, G., Dragomir, A., Woodard, J., Fang, F., Schwartz, T., Abbate, L., Callahan, L., et al. (2007). Prevalence of knee symptoms and radiographic and symptomatic knee osteoarthritis in African Americans and Caucasians: the Johnston County Osteoarthritis Project. *The Journal of Rheumatology*, 34(1):172–180.
- Kaufman, K., Hughes, C., Morrey, B., Morrey, M., and An, K. (2001). Gait characteristics of patients with knee osteoarthritis. *Journal of Biomechanics*, 34(7):907–915.

- Kellgren, J. and Lawrence, J. (1957). Radiological assessment of osteo-arthrosis. *Annals of the Rheumatic Diseases*, 16(4):494–502.
- Kirkley, A., Webster-Bogaert, S., Litchfield, R., Amendola, A., MacDonald, S., McCalden, R., and Fowler, P. (1999). The effect of bracing on varus gonarthrosis. *Journal of Bone and Joint Surgery (American)*, 81(4):539–548.
- Komistek, R., Dennis, D., Northcut, E., Wood, A., Parker, A., and Traina, S. (1999). An in vivo analysis of the effectiveness of the osteoarthritic knee brace during heel-strike of gait. *The Journal of Arthroplasty*, 14(6):738–742.
- Kuroyanagi, Y., Nagura, T., Kiriya, Y., Matsumoto, H., Otani, T., Toyama, Y., and Suda, Y. (2011). A quantitative assessment of varus thrust in patients with medial knee osteoarthritis. *The Knee*, Epub ahead of print.
- Kurtz, S., Ong, K., Lau, E., Mowat, F., and Halpern, M. (2007). Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. *The Journal of Bone and Joint Surgery (American)*, 89(4):780–785.
- Landry, S., McKean, K., Hubley-Kozey, C., Stanish, W., and Deluzio, K. (2007). Knee biomechanics of moderate OA patients measured during gait at a self-selected and fast walking speed. *Journal of Biomechanics*, 40(8):1754–1761.
- Lane, N., Brandt, K., Hawker, G., Peeva, E., Schreyer, E., Tsuji, W., and Hochberg, M. (2011). OARSI-FDA initiative: defining the disease state of osteoarthritis. *Osteoarthritis and Cartilage*, 19(5):478–482.
- Lindenfeld, T., Hewett, T., and Andriacchi, T. (1997). Joint loading with valgus bracing in patients with varus gonarthrosis. *Clinical Orthopaedics and Related Research*, Nov(344):290–297.
- Maly, M. (2008). Abnormal and cumulative loading in knee osteoarthritis. *Current Opinion in Rheumatology*, 20(5):547–552.
- Maquet, P. (1980). The biomechanics of the knee and surgical possibilities of healing osteoarthritic knee joints. *Clinical Orthopaedics and Related Research*, 146:102–110.
- Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., and Shimada, S. (2002). Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. *Annals of the Rheumatic Diseases*, 61(7):617–622.
- Mündermann, A., Dyrby, C., and Andriacchi, T. (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. *Arthritis & Rheumatism*, 52(9):2835–2844.
- Neogi, T., Jordan, J., and Zhang, Y. (2011). OARSI Primer: Epidemiology of OA. <http://primer.oarsi.org/>.

- Newell, R., Hubley-Kozey, C., Stanish, W., and Deluzio, K. (2008). Detecting differences between asymptomatic and osteoarthritic gait is influenced by changing the knee adduction moment model. *Gait & Posture*, 27(3):485–492.
- Otis, J., Backus, S., Campbell, D., Furman, G., Montalvo, E., Warren, R., Wickiewicz, T., and Peterson, M. (2000). Valgus bracing for knee osteoarthritis: A biomechanical and clinical outcome study. In *46th Annual Meeting: Orthopaedic Research Society*, page 0262.
- Otis, J., Backus, S., Pollo, F., Wickiewicz, T., and Warren, R. (1996). Load sharing at the knee during valgus bracing for medial compartment osteoarthritis. *Gait & Posture*, 4(2):189.
- Otten, E. (2003). Inverse and forward dynamics: models of multi-body systems. *Philosophical Transactions of the Royal Society of London. Series B: Biological Sciences*, 358(1437):1493–1500.
- Piazza, S. and Cavanagh, P. (2000). Measurement of the screw-home motion of the knee is sensitive to errors in axis alignment. *Journal of Biomechanics*, 33(8):1029–1034.
- Pollo, F., Otis, J., Backus, S., Warren, R., and Wickiewicz, T. (2002). Reduction of medial compartment loads with valgus bracing of the osteoarthritic knee. *The American Journal of Sports Medicine*, 30(3):414–421.
- Public Health Agency of Canada (2010). Life with Arthritis in Canada: A personal and public health challenge. <http://www.phac-aspc.gc.ca/cd-mc/arthritis-arthrite/lwaic-vaaac-10/index-eng.php/>.
- Ramsey, D., Briem, K., Axe, M., and Snyder-Mackler, L. (2007). A mechanical theory for the effectiveness of bracing for medial compartment osteoarthritis of the knee. *Journal of Bone and Joint Surgery (American)*, 89(11):2398–2407.
- Richards, J., Sanchez-Ballester, J., Jones, R., Darke, N., and Livingstone, B. (2005). A comparison of knee braces during walking for the treatment of osteoarthritis of the medial compartment of the knee. *Journal of Bone and Joint Surgery (British)*, 87(7):937–939.
- Rijkelijkhuizen, J., Meijer, H., Baan, G., and Huijting, P. (2007). Myofascial force transmission also occurs between antagonistic muscles located within opposite compartments of the rat lower hind limb. *Journal of Electromyography and Kinesiology*, 17(6):690–697.
- Roemhildt, M., Coughlin, K., Peura, G., Badger, G., Churchill, D., Fleming, B., and Beynon, B. (2010). Effects of increased chronic loading on articular cartilage material properties in the lapine tibio-femoral joint. *Journal of Biomechanics*, 43(12):2301–2308.
- Sale, J., Gignac, M., and Hawker, G. (2008). The relationship between disease symptoms, life events, coping and treatment, and depression among older adults with osteoarthritis. *Journal of Rheumatology*, 35(2):335–342.

- Schipplein, O. and Andriacchi, T. (1991). Interaction between active and passive knee stabilizers during level walking. *Journal of Orthopaedic Research*, 9(1):113–119.
- Schnitzer, T., Popovich, J., Andersson, G., and Andriacchi, T. (1993). Effect of piroxicam on gait in patients with osteoarthritis of the knee. *Arthritis & Rheumatism*, 36(9):1207–1213.
- Self, B., Greenwald, R., and Pflaster, D. (2000). A biomechanical analysis of a medial unloading brace for osteoarthritis in the knee. *Arthritis Care & Research*, 13(4):191–197.
- Sharma, L., Hurwitz, D., Thonar, E., Sum, J., Lenz, M., Dunlop, D., Schnitzer, T., Kirwan-Mellis, G., and Andriacchi, T. (1998). Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. *Arthritis & Rheumatism*, 41(7):1233–1240.
- Takahashi, T., Yoshihara, E., Mukai, K., Ohmura, H., and Hiraga, A. (2010). Use of an implantable transducer to measure force in the superficial digital flexor tendon in horses at walk, trot and canter on a treadmill. *Equine Veterinary Journal*, 42:496–501.
- Tallon, D., Chard, J., and Dieppe, P. (2000). Exploring the priorities of patients with osteoarthritis of the knee. *Arthritis Care & Research*, 13(5):312–319.
- Taylor, W., Heller, M., Bergmann, G., and Duda, G. (2004). Tibio-femoral loading during human gait and stair climbing. *Journal of Orthopaedic Research*, 22(3):625–632.
- The Arthritis Society of Canada (2004). Arthroscopy. <http://www.arthritis.ca/>.
- Thorp, L., Sumner, D., Block, J., Moio, K., Shott, S., and Wimmer, M. (2006). Knee joint loading differs in individuals with mild compared with moderate medial knee osteoarthritis. *Arthritis & Rheumatism*, 54(12):3842–3849.
- Thorp, L., Sumner, D., Wimmer, M., and Block, J. (2007). Relationship between pain and medial knee joint loading in mild radiographic knee osteoarthritis. *Arthritis Care & Research*, 57(7):1254–1260.
- Toriyama, M., Deie, M., Shimada, N., Otani, T., Shidahara, H., Maejima, H., Moriyama, H., Shibuya, H., Okuhara, A., and Ochi, M. (2011). Effects of unloading bracing on knee and hip joints for patients with medial compartment knee osteoarthritis. *Clinical Biomechanics*, 26(5):497–503.
- Vaughan, C., Davis, B., and Jeremy, C. (1992). *Dynamics of human gait*. Human Kinetics Publishers, Champaign, Illinois.
- Winby, C., Lloyd, D., Besier, T., and Kirk, T. (2009). Muscle and external load contribution to knee joint contact loads during normal gait. *Journal of Biomechanics*, 42(14):2294–2300.

- Yelin, E., Cisternas, M., Pasta, D., Trupin, L., Murphy, L., and Helmick, C. (2004). Medical care expenditures and earnings losses of persons with arthritis and other rheumatic conditions in the United States in 1997: total and incremental estimates. *Arthritis & Rheumatism*, 50(7):2317–2326.
- Zhang, W., Moskowitz, R., Nuki, G., Abramson, S., Altman, R., Arden, N., Bierma-Zeinstra, S., Brandt, K., Croft, P., Doherty, M., et al. (2008). OARSI recommendations for the management of hip and knee osteoarthritis, Part II: OARSI evidence-based, expert consensus guidelines. *Osteoarthritis and Cartilage*, 16(2):137–162.
- Zhao, D., Banks, S., Mitchell, K., D’Lima, D., Colwell, C., and Fregly, B. (2007). Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *Journal of Orthopaedic Research*, 25(6):789–797.