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Walk Softly and Carry a Big Stick: Strategies to Decrease Dynamic Knee Joint Loading

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A thesis submitted in partial fulfillment of the requirements for the Doctor of Philosophy degree in Health and Rehabilitation Sciences

WALK SOFTLY AND CARRY A BIG STICK:
STRATEGIES TO DECREASE DYNAMIC KNEE JOINT LOADING

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Abstract

Excessive dynamic loading of the knee joint, quantified indirectly during three dimensional gait analysis, is a risk factor for the progression of knee osteoarthritis (OA). The overall objective of this thesis was to explore the effects of prolonged walking and the use of Nordic walking poles on selected gait characteristics indicative of knee joint load. The first study evaluated the time-varying behaviour, reliability, and validity of selected gait kinematics during 60 minutes of treadmill walking in 20 healthy adults. Maximum lateral trunk lean angle and maximum toe-out angle did not change over time, were consistent from day to day and were consistent with values assessed during over-ground gait analysis, suggesting that these measures are appropriate for use in studying potential adaptive gait mechanisms. The second study compared the time-varying behaviour of selected gait kinematics during 30 minutes of treadmill walking in 20 participants with, and 20 participants without, medial compartment knee OA, and explored correlations between these gait kinematics and pain intensity. Trunk lean, toe-out, and pelvic rise were different between those with and without knee OA, but did not systematically change over time in either group. Trunk lean and contralateral pelvic drop were significantly correlated to pain intensity. The third study was a technical report describing the use of three dimensional gait analysis and a Nordic walking pole instrumented with a compression load cell. This methodology was then used in the fourth study to evaluate the effect of walking poles used by 34 patients with medial compartment knee OA. Despite small reductions in the vertical ground reaction force, walking with poles increased the frontal plane lever arm, and therefore the knee adduction moment. The pole force in the vertical direction was inversely related to the increase in first peak knee adduction moment. Overall, this thesis suggests that, although biomechanically plausible, patients with knee OA do not naturally adopt gait characteristics or use walking poles in a way to decrease knee joint loads. These findings support the need for future efforts directed at explicitly teaching walking techniques, including the optimal use of assistive devices, to decrease knee joint loading.

Keywords: compensatory gait biomechanics, human locomotion, measurement properties, gait compensation, pain, pelvic tilt, prolonged gait, knee adduction moment, lever arm, hiking poles, assistive walking device and knee joint loading

Co-Authorship Statement

This thesis contains materials from one published manuscript (Chapter 2), one manuscript accepted for publication (Chapter 3), and two manuscripts in preparation to be submitted for peer review (Chapter 4 and 5). Daniel Bechard was primarily responsible for collection, analysis, and interpretation of data for all studies in this thesis. Dr. Trevor Birmingham and Dr. Thomas Jenkyn were co-authors for all four studies (Chapter 2 to 5) and assisted in study design, data analysis, and manuscript preparation. Dr. Aleksandra Zecevic was co-author for Chapters 1, 2, 3, and 5 and assisted in study design, and manuscript preparation. Dr. Robert Giffin and Mr. Ian Jones were co-authors on Chapters 3 and 5. They assisted in study design, patient recruitment, and data collection. Ms. Kristyn Leitch was a co-author for Chapter 5 and assisted with data processing and manuscript writing. Daniel Bechard was the sole author of Chapter 6.

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List of Abbreviations, Symbols, Nomenclature

ANOVA – analysis of variance

AP - Anteroposterior

BMI – body mass index

BW – body weight

CI – confidence interval

GRF – ground reaction force

ICC – intraclass correlation coefficient

KAM – knee adduction moment

KL Grade – Kellgren and Lawrence grade of OA severity

MAA – mechanical axis angle

ML - mediolateral

OA – osteoarthritis

SD – standard deviation

SEM – standard error of measurement

%BW*Ht – Percent body weight times height

Chapter 1:

Introduction and Background

1.1 Overview

Excessive dynamic loading of the knee joint, quantified indirectly during three dimensional gait analysis, is a risk factor for the progression of knee osteoarthritis (OA) – a leading cause of physical disability and health care use. The purpose of the present chapter was to provide an introduction to knee OA, gait analysis, and knee joint loading. Selected gait patterns (i.e. kinematics) and interventions (i.e. therapeutic devices such as canes and walking poles) proposed to reduce knee joint loading were described to provide relevant background and rationale for the studies in this thesis.

1.2 Articular Cartilage

Articular cartilage can be described as a thin, dense, connective tissue that covers the epiphyses of synovial joints. This tissue has two main functions, first, to better distribute loads across weight bearing surfaces and secondly, to reduce friction and wear that is associated with moving loaded joint surfaces.

The structural components of articular cartilage are chondrocytes, collagen, proteoglycans, and water. Chondrocytes are small in population, but responsible for the maintenance of the extracellular matrix within cartilage. Within the matrix, collagen provides tensile strength, while proteoglycans retain water and other ions. A healthy, inflated structure is maintained through proper function of these components that is adaptable to dynamic motion and variable loads through the movement of extracellular fluid.

The viscoelastic properties of articular cartilage allow tissue to adapt during gait through a ‘creep’ and ‘stress relaxation’ response to loading. In a ‘creep’ response, interstitial fluid will exude from the tissue when a constant load is applied. This will cause the tissue to deform in a first rapid response that will slow until solid matrix take up a sufficient amount of the load. When equilibrium is achieved the tissue deformation will then stabilize. In a ‘stress relaxation’ response, a stress is applied in which fluid exudation takes place. After a maximum strain, deformation due to fluid exiting the tissue occurs, much like in the creep response. Remaining fluid will then redistribute within the tissue allowing a relaxation of the tissue until equilibrium is achieved (Mow & Hung, 2001). Through proper function of these principles, healthy cartilage can be maintained, however damage to the cartilage can result from high or rapid loads leading to a ‘flushing’ of essential proteoglycan and collagen compounds. OA occurs, when the repair cannot keep up with the destructive process.

1.3 Osteoarthritis

Osteoarthritis is a progressive, degenerative condition of the articular cartilage that involves a complex interaction between biochemical and biomechanical pathologies (Lawrence et al., 2008). OA usually occurs in weight bearing joints, of which the medial compartment of the knee is most commonly affected. The onset of OA is thought to be triggered by progressive ‘wear and tear’ due to cyclic joint loading during regular, everyday activity. However, OA can also be initiated through a traumatic event such as an anterior cruciate ligament tear which compromises joint stability (Poole, 1999). In each case, the damage of the articular cartilage exceeds the repairing capacity. Risk factors for OA development include genetic predisposition, age, sex, obesity, and previous joint injury. Women tend to have a higher prevalence and severity of knee OA; however, symptoms present similarly in both sexes (O'Connor, 2006).

Initially arthritis is identified through symptoms such as weight bearing pain, crepitus (cracking or popping sounds or sensations), stiffness, and loss of range of motion that contribute to activity limitations and compensations (Altman et al., 1986; Hunt et al., 2008; Mundermann, Dyrby, & Andriacchi, 2005). Diagnosis is usually completed by radiograph in conjunction with Altman's criteria (1986) and grading of severity through the Kellgren and Lawrence scale (1957). A positive assessment through Altman's criteria includes knee pain, age greater than 18 years, radiographic evidence of osteophytes (bone spurs), and at least one of: age greater than 50 years, morning knee stiffness lasting longer than 30 minutes, or crepitus associated with motion of the knee. Kellgren and Lawrence devised a commonly used classification system in order to rate the severity of disease on a five point scale. No indication of OA is represented by zero while severe OA which includes joint space narrowing, presence of osteophytes and severe subchondral sclerosis (thickening of bone under the cartilage) is given a rating of four. Individuals often seek care due to pain attributing to activity limitations; however, pain is not the deciding factor in a positive OA diagnosis.

1.4 The Impact of Osteoarthritis

Conditions, such as OA, where acute and long term activity limitations are prevalent are costly in today's society. Sixteen percent of Canadians over the age of 15 (Health Canada, 2003) and 21.5% of Americans over 18 (Bitton, 2009) are affected by arthritis, of which OA is the most common form. This is projected to increase to 21% by 2026 for Canadians and 25% of Americans by 2030, primarily due to the rapidly increasing rate of obesity and aging of the population. The cost of arthritis in developed nations is staggering. In 1997, costs of arthritis were estimated to be between 1 and 2.5% of the gross domestic product of five nations including the US and Canada (Reginster, 2002). During this year, the total medical expenditures for arthritis and other rheumatic disabilities in the US was $233.5 billion, but by 2003, costs had increased to $321.8 billion (Yelin, 2007).

The individual impact is equally significant. In Canada, 50% of those with arthritis under the age of 75 report limitations in activities that take place at home, work, school, or other settings. Of those with limitations, 40% require assistance with activities of daily living (Health Canada, 2003). Individual costs that result from these limitations range from increased care required, more days spent seeking medical treatment, to lost wages (Bitton, 2009).

Depending on the severity of symptoms, the individual will reassess daily activities and community participation in an incremental process. *Valued activities* are the first to be reassessed and modified when symptoms begin. *Community mobility, household activity, in-home mobility,* and *personal care* follow as severity increases. As the severity of the symptoms progress, independence is continuously challenged (Cott & Gignac, 1999). Individuals with conditions that affect their mobility will initially modify *valued activities* such as sport and recreation. If a task cannot be completed with a perception of safety and confidence, participation is likely reduced or stopped completely as a final resort (Maly & Krupa, 2007). *Community mobility* is altered next. It is described as doing what you want when you want (leaving the home to go shopping, or driving a car). This stage could result in avoidance of leaving home due to the perceived risk to safety, and pain (Charmez, 1995). As severity of symptoms progress, independence in *household activities* is threatened (general home care and meal preparation). Individuals will modify their daily tasks in order to become more time and effort efficient because of increasing pain and mobility disruption. At this stage, the attention is only given to the basic activities needed to take care of themselves and their household. When *in-home mobility* is challenged, tasks such as walking from room to room and up stairs can be threatened. Finally, the most severe limitation affects *personal care.* This includes difficulty or inability to complete fundamental daily tasks such as dressing, shaving, and bathing (Charmez, 1995; Cott & Gignac, 1999; Maly & Krupa, 2007).

At each stage of reassessment, symptoms of the condition provide reason for a perceived risk of pain, further damage, and inability to complete otherwise standard tasks, therefore these activities are constantly modified or removed from daily life. Individuals that live with knee OA experience increasing frustration of limitations to independence. This frustration often leads to depression, decreasing self efficacy, and attempts to engage in activities that risk pain, personal safety, and further joint damage (Bourret, Bernick, Cott, & Kontos, 2002; Maly & Krupa, 2007). With limited independence, compromised safety, and frustration, coupled with increased mobility restrictions, a sedentary lifestyle and social isolation become an increasing risk to general well being. For these reasons, research that analyses the causes of disease initiation and progression and offers deeper insight into potential solutions is essential.

1.5 3-D Gait Analysis and the Knee Adduction Moment

Today, 3D motion capture systems are frequently used to examine human mobility. Most systems are based on a system of multiple high resolution digital cameras that identify and record the positions of reflective markers placed on the skin of the patient, and force plates that detect ground reaction forces acting on the body in three dimensions.

Traditionally, these systems are used for analysis of over-ground gait. The treadmill however provides the advantage of removing the constraints of the traditional over-ground walkway allowing for a longer, uninterrupted walking path. Despite its advantages, treadmill walking is different from over-ground walking (Marsh et al., 2006; Matsas, Taylor, & McBurney, 2000; Riley, Paolini, Croce, Paylo, & Kerrigan, 2006; Wass, Taylor, & Matsas, 2005). This may be a result of the increased level of coordination required and difference in visual perception during gait. These differences may also be exacerbated depending on the level of experience, and age of the individual. Differences can however, be mitigated through proper familiarization techniques (Matsas et al., 2000; Wass et al., 2005). Walking that occurs on a treadmill could therefore better mimic the exertion, fatigue, and activity onset pain that is characteristic of daily activity.

Regardless of any drawbacks, use of the treadmill provides a suitable method for measuring gait kinematics over prolonged periods of time.

Whether over-ground or on a treadmill, 3D motion capture systems do not directly measure internal joint forces, however inverse dynamics calculations enable accurate estimates of forces acting on joints (Winter, 1990). It has been proposed that interventions anchored on objective, accurate data that motion capture provides can lead to more successful treatment resulting in fewer hospital visits for secondary invasive procedures and faster rehabilitation resulting in a greater quality of life (Hailey, & Tomie, 2000; Wren, Woolf, & Kay, 2005).

Investigation of knee joint load in patients with knee OA is an example of how motion capture can reveal the mechanisms that contribute to mobility limitations (McGibbon & Krebs, 2002; Andriacchi et al., 2004; Mundermann et al., 2004; Sharma et al., 2008; Rutherford, Hubley-Kozey, Deluzio, & Stanish, 2008; Ramsey, Snyder-Mackler, Lewek, Newcomb, & Rudolph, 2007). Knee joint load is commonly described using the external knee adduction moment. This moment is composed of the frontal plane ground reaction force acting on the body, lever arm that extends from the ground reaction force line of action to the frontal plane centre of rotation of the knee, and inertial properties of the lower limb (Figure 1.1 A). During gait, the line of action from the ground reaction force passes medially to the centre of rotation of the knee. This creates a torque tending to adduct the tibia with respect to the femur. The knee adduction moment usually follows a ‘double hump’ pattern, with first peak occurring in the first half of stance while second peak occurs in the second half of stance. The knee adduction moment has been found to have a positive correlation with compressive forces in the medial compartment (Schipplein & Andriacchi, 1991) and is considered a valid (Hurwitz, Sumner, Case, Block, & Andriacchi, 1998; Jackson et al., 2004; Thorp et al., 2006; Wada et al., 2001) and reliable (Birmingham, Hunt, Jones, Jenkyn, & Giffin, 2008) proxy for knee joint loading of the medial compartment.

Mechanical axis angle continues to be the best predictor of the knee adduction moment (Hurwitz, Ryals, Case, Block, & Andriacchi, 2002; Hunt et al., 2008) and consists of a line drawn in the frontal plane from the centre of the ankle and centre of the femoral head to the centre of the knee (Figure 1.1 B). This creates an angle of deviation that represents a quantification of alignment. Individuals with neutral (mechanical axis angle) alignment experience 75% of the load passing through their knee in the medial compartment (Hsu, Himeno, Coventry, & Chao, 1990). Individuals with valgus alignment (positive mechanical axis angle) will experience a greater than normal load on the lateral compartment, whereas more common in individuals with knee OA a varus alignment (negative mechanical axis angle) will result in increased loading on the medial compartment. High knee adduction moments are considered a major risk factor for the initiation, severity, and progressions of OA. Miyazaki et al. (2002) showed that an increase in the peak knee adduction moment of 1% body weight times height (BW*Ht) (approximately 25% increase of absolute load) resulted in a 6.5 fold greater risk of progression of radiographic knee OA over a six year period, increasing the risk of further mobility limitations. The altering of joint biomechanics therefore has the potential to accelerate the degeneration of articular cartilage. This presents a cyclic process for patients with medial compartment knee OA. As cartilage degeneration and joint space narrowing occurs, malalignment increases and leads to increased loading thereafter (Figure 1.2).

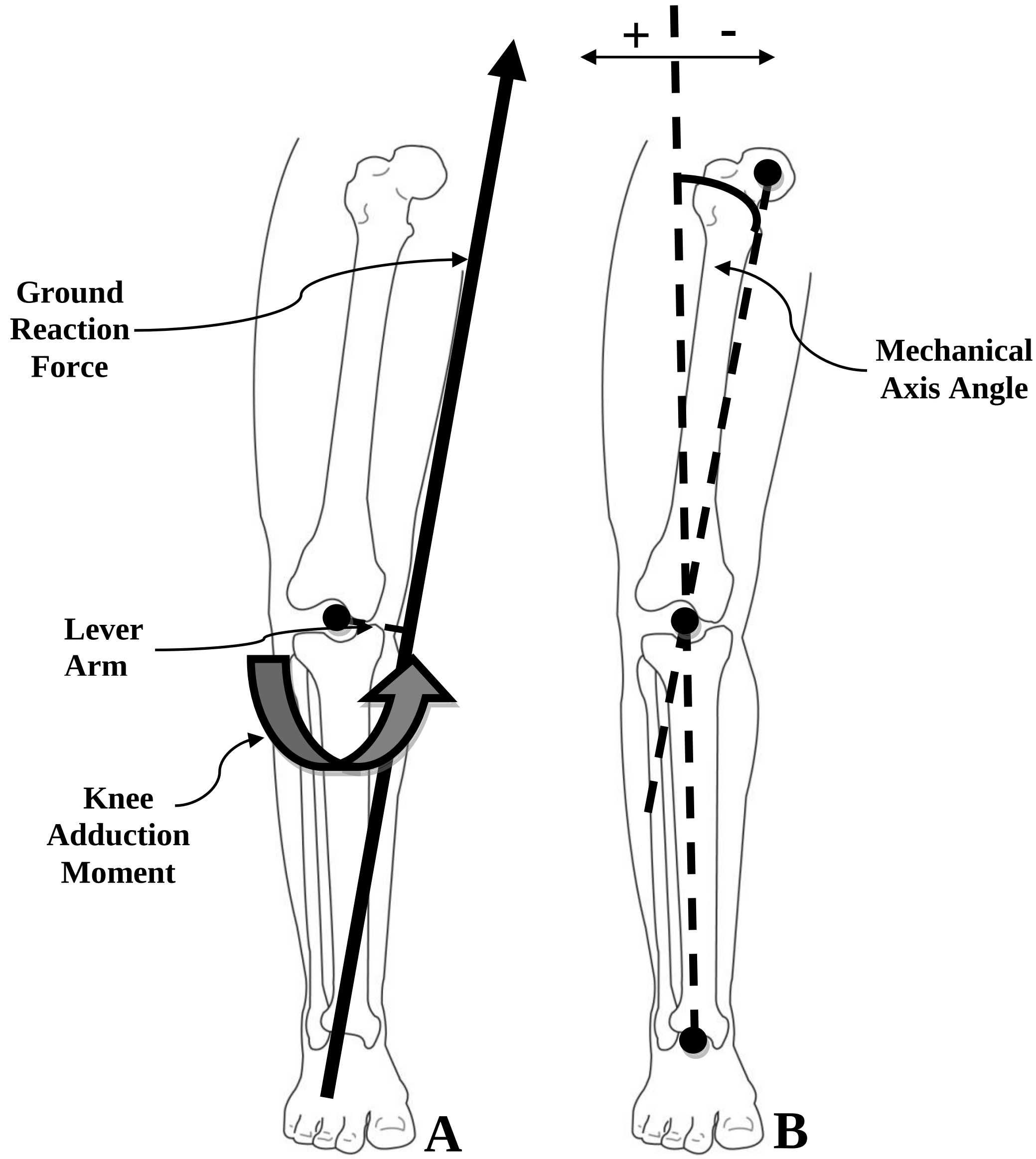

Figure 1.1. An illustration of the knee adduction moment (A) and mechanical axis angle (B). The knee adduction moment consists of the line of action of the ground reaction force and the lever arm extending from the centre of the knee to the line of the ground reaction force. The mechanical axis angle is a measure of limb alignment and consists of the angle created by lines drawn from the centre of the femoral head and centre of the ankle to the centre of the knee. Negative values indicate varus alignment (bow legs) and positive values indicate valgus alignment (knock knees).

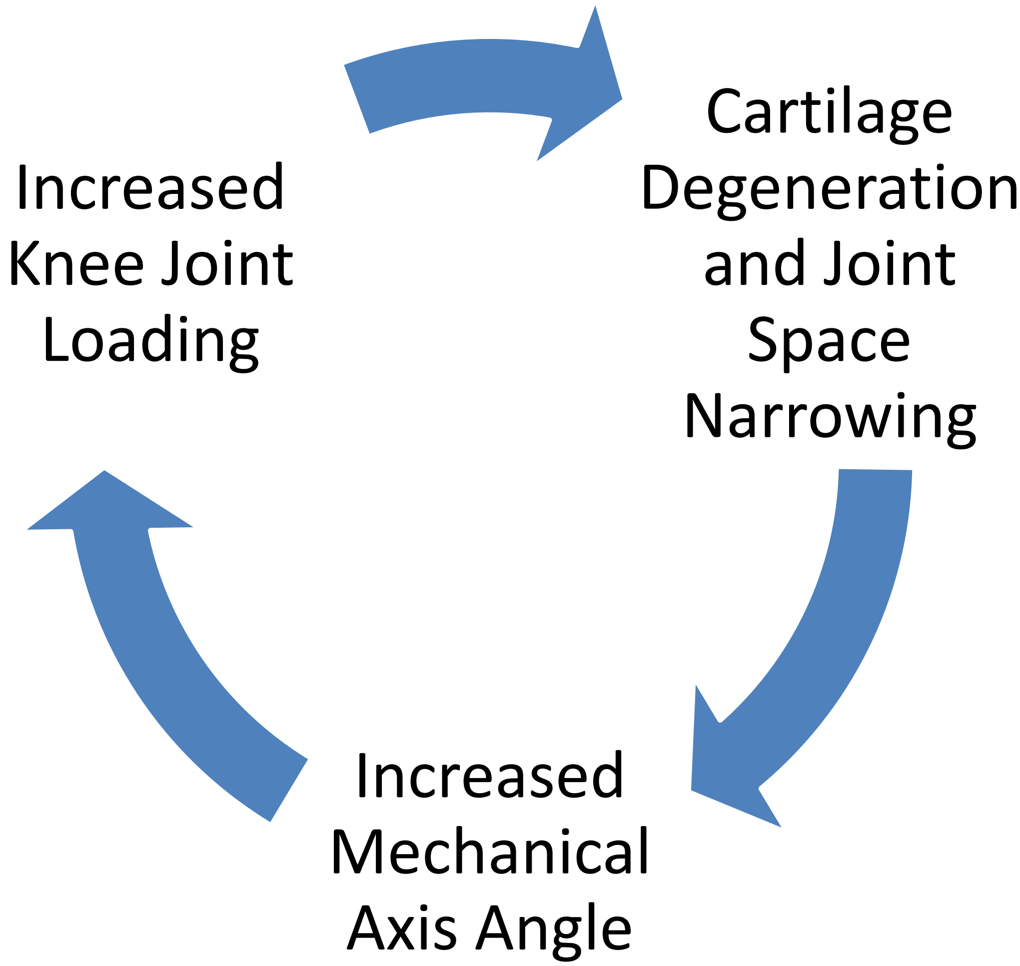

Figure 1.2. A schematic diagram of the degenerative cycle that occurs in patients with knee osteoarthritis that includes a cyclic process of cartilage degeneration and joint space narrowing, increased mechanical axis angle, and increased knee joint load.

1.6 Methods of Reducing the Knee Adduction Moment

The primary goal of any treatment for symptomatic knee OA is to increase function and reduce pain. Patients with knee OA are encouraged to participate in physical activity, yet the potential for further degeneration and increased knee loading is considerable. However, strategies can be developed to mitigate this potential.

During quiet standing the mass of the individual is shared between limbs, however during gait, this force regularly reaches 1.2 times bodyweight acting through a single limb. The line of action of the ground reaction force acting further away from the centre of rotation of the knee causes a greater knee adduction moment. A common strategy to address the symptoms of knee OA has focused on the reduction of the knee adduction moment and the transfer of a portion of load, borne by the medial compartment, to the lateral. A well established invasive treatment option for those with unicompartmental knee OA is a medial opening wedge high tibial osteotomy (HTO). The purpose of this surgery is to realign the affected limb from a varus alignment to neutral (mechanical axis approximately zero). By moving the shank laterally, the intention is to move the ground reaction force vector closer to the knee's centre of rotation during the stance phase of gait thereby reducing the knee adduction moment. This redistributes the load travelling through the knee, thereby relieving the affected compartment (Giffin & Shannon, 2007). This corrective surgery is designed for those with varus alignment and unicompartmental knee OA (genu varum) and therefore, tolerate a higher than normal load in their medial compartment. Current short term studies show the positive effects of this invasive technique that result in lower knee adduction moment, improved quality of life and daily function (Birmingham et al., 2009).

Invasive treatment strategies are often considered a last resort. The more common methods by which individuals manage the symptoms associated with knee OA are through conservative techniques such as corrective devices. Two common non-invasive methods of reducing knee adduction moment are medial unloader braces and lateral heel wedges.

Several exhaustive literature reviews advocate the benefits of medial unloader braces for those with knee OA (Gravlee & Van Durme, 2007; Gross & Hillstrom, 2008; Pollo & Jackson, 2006; Ramsey, Briem, Axe, & Snyder-Mackler 2007). The purpose of the medial unloader brace is to correct malalignment and improve medial lateral stability in patients with medial compartment knee OA. Unloader braces improve alignment during quiet standing and maintain that alignment during gait (Komistek et al., 1999). These devices have been shown to reduce the knee adduction moment from between 10 to 13% (Lindenfeld, Hewett, & Andriacchi, 1997; Pollo, Otis, Backus, Warren, & Wickiewicz, 2002) while increasing function and decreasing levels of pain during daily activity (Gaasbeek, Groen, Hampsink, van Heerwaarden, & Duysens, 2007; Kirkley, Webster-Bogaert, & Litchfield, 1999).

The use of a lateral heel wedge is another non-invasive method in the treatment of knee OA. The lateral heel wedge adjusts the position of stance thereby changing the location of the centre of pressure of the affected limb. Limited but growing evidence exists regarding the efficacy of these devices as therapeutic intervention. Kerrigan et al.

(2002) and Crenshaw, Pollo, and Calton (2000) used lateral wedges between 5 and 10 degrees and reported reductions of less than 10% knee adduction moment, however the effects on pain and function are currently mixed. Maillefert et al. (2001) and Baker et al. (2007) reported no difference in pain or function scores whereas other studies (Rodrigues et al. 2008; Rubin & Menz, 2005) advocate the device's role in improving these same variables.

1.6.1 Trunk lean and Toe-out

Various gait kinematics, identified using 3-D motion analysis, have been associated with decreased knee joint load during gait. In particular, increased lateral trunk lean over the stance limb (Andriacchi & Mundermann, 2006; Hunt et al., 2008; Tanaka et al., 2008; Mundermann et al., 2005; Mundermann, Asay, Mundermann, & Andriacchi, 2008) and increased toe-out (foot progression) angle (Andrews, Noyes, Hewett, & Andriacchi, 1996; Chang et al. 2007; Guo, Axe, & Manal, 2007; Hurwitz et al., 2002; Jenkyn, Hunt, Jones, Giffin, & Birmingham, 2008; Rutherford et al., 2008) have been consistently reported to decrease the knee adduction moment. Hunt et al. (2008) investigated the role of gait kinematics in the variation of the knee adduction moment in 120 patients with knee OA. They determined that trunk lean and toe out explained 13 and 12% of the variation in knee adduction moment respectively. These gait kinematics have been frequently discussed as potential adaptive gait patterns adopted by patients with knee OA in an attempt to lessen the load and symptoms on the affected medial compartment (Andrews et al., 1996; Andriacchi & Mundermann, 2006; Hunt et al., 2008; Hurwitz et al. 2002; Mundermann et al. 2004; Mundermann et al., 2005; Tanaka et al., 2008). Lateral trunk lean and toe-out have also been discussed as a possible therapeutic intervention for patients with knee OA (Chang et al., 2007; Gou et al., 2007; Mundermann et al., 2008).

Deviating the trunk towards the stance limb reduces the knee adduction moment by moving the line of action of the ground reaction force closer to the centre of rotation of the knee, thereby decreasing the frontal plane lever arm and knee adduction moment.

Previous investigations have focused on the role of lateral trunk lean in both healthy adults (Mundermann et al., 2008) and patients with knee OA (Tanaka et al., 2008; Hunt et al., 2008; Hunt, Wrigley, Hinman & Bennell, 2010). Mundermann et al. (2008) investigated the effects of increased lateral trunk lean on the knee adduction moment in 19 healthy adults. Increasing trunk lean by 10 degrees resulted in a reduction of knee adduction moment by 65% on average (Mundermann et al., 2008).

In patients with knee OA, trunk lean angles of two to five degrees are more common nevertheless, the effects are still apparent (Andriacchi & Mundermann, 2006; Hunt et al., 2008; Mundermann et al., 2005; Tanaka et al., 2008;). For example, Hunt et al. (2010) investigated proximal segment walking mechanics in 75 patients with knee OA of varying severities and 20 healthy adults. They reported an average trunk lean angle of 5.0 degrees in patients with severe knee OA compared to 1.6 degrees in individuals with no knee pain.

This thesis defines lateral trunk lean as the angle of a line drawn from the midpoint of the anterior superior iliac spines (ASISs) to the midpoint of the anterior tips of the acromion processes with respect to the vertical (Figure 1.3). Positive values indicate a trunk lean towards the stance limb while a negative value indicates a lean towards the swing limb.

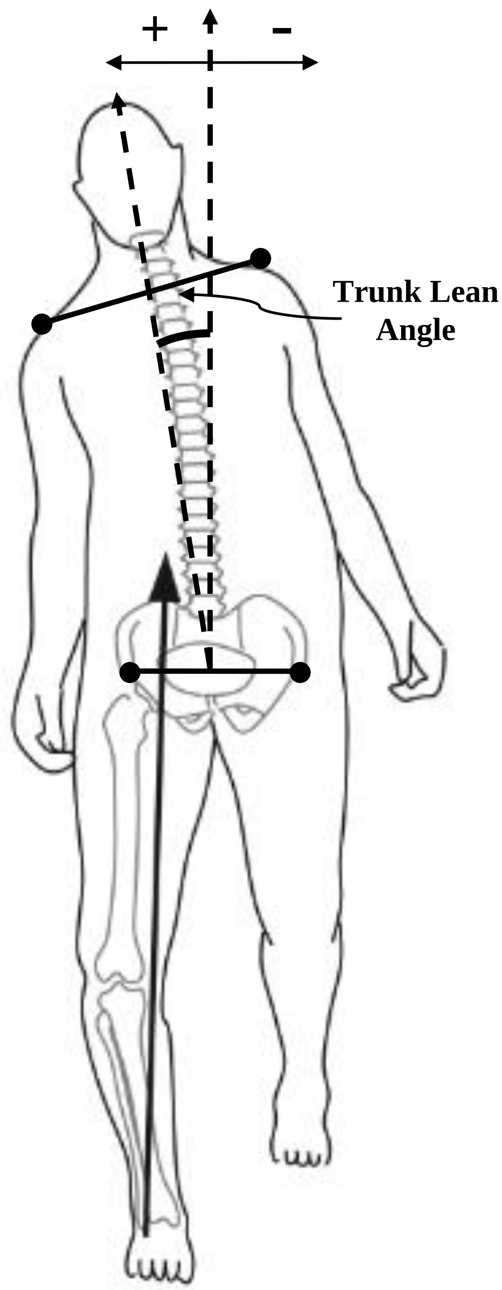

Figure 1.3. An illustration of trunk lean angle measured from the midpoint of a line connecting the anterior superior iliac spines and the midpoint of the line connecting the acromion processes with respect to the vertical. Frontal plane ground reaction force is also shown. Positive values represent a lean towards the stance limb while negative values represent a lean towards the swing limb.

Toeing-out during gait is also a commonly proposed compensation strategy suggested to reduce knee joint loads (Andrews et al., 1996; Chang et al., 2007; Guo et al., 2007; Hurwitz et al., 2002; Jenkyn et al., 2008; Rutherford et al., 2008). Due to the centre of pressure following the lateral border of the foot throughout stance, decreases in the knee's frontal plane lever arm and adduction moment are most pronounced during later stance (Hurwitz et al., 2002). Variability for toe-out is quite high (-2.2 degrees toe in (Jenkyn et al., 2008) to almost 40 degrees toe-out (Chang et al., 2007)), however toe-out has been consistently shown to have a negative correlation with the knee adduction moment in both patients with knee OA (Chang et al., 2007; Gou et al., 2007; Hurwitz et al., 2002; Jenkyn et al., 2008; Rutherford et al., 2008) and healthy adults (Andrews et al., 1996; Teichtahl, Cicuttini, Janakiramanan, Davis, & Wluka, 2006).

The effectiveness of reducing the knee adduction moment through increases in toe-out has been demonstrated in the past. Gou et al. (2007) investigated voluntary implementation of increased toe-out angle during gait in 10 patients with knee OA. An average increase in toe-out angle of 16.6 degrees resulted in a 38% reduction in second peak knee adduction moment. This is of importance since a smaller toe-out angle has been associated with a greater likelihood of OA progression after 18 months (Chang et al., 2007). The toe-out angle has been defined in this thesis as the angle between a line drawn from the centre of the ankle to the head of the second metatarsal and the forward progression of the body (Figure 1.4). Positive values indicate a toe-out while negative values indicate a toe-in.

In summary, although lateral trunk lean and toe-out clearly affect knee joint loading, and disease progression, their roles as adaptive, compensatory mechanisms are currently unclear. Specifically, it is not known if magnitudes of trunk lean and toe-out change as a response to prolonged bouts of gait, or activity onset pain.

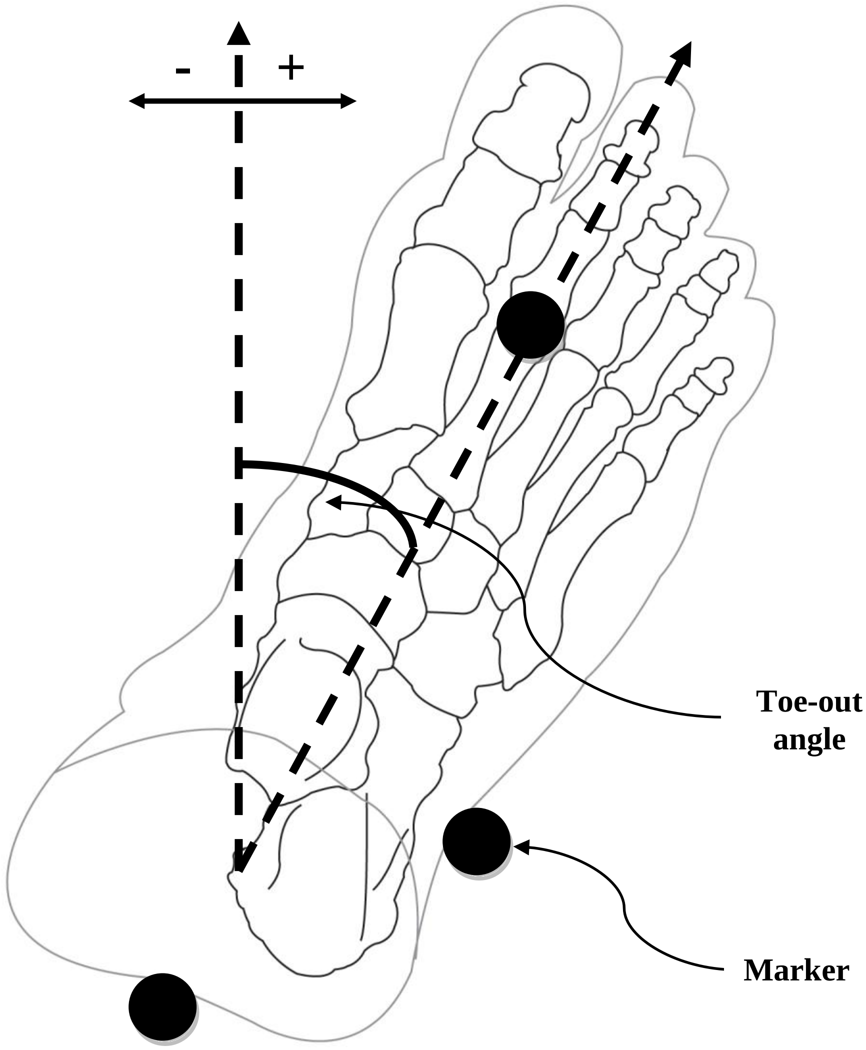

Figure 1.4. An illustration of toe-out angle measured as a line connecting the middle of the ankle to the head of the second metatarsal with respect to the forward progression of the body. Positive values indicate toeing out while negative values indicate toeing in.

Table 1.1: Summary of findings from articles investigating trunk lean

Authors	N (M/F)	Participants	Age	Primary Findings
Hunt et al. (2011)	9(3/6)	Healthy	18.6±0.7	Trunk lean of 4, 8, and 12 degrees was associated with 7, 21, and 25% reductions in knee adduction moment.
Hunt et al. (2010)	95(53/42)	25 Severe 25 Moderate 25 Mild 20 No knee pain	68.0±6.6 63.6±8.4 61.2±7.7 63.2±12.4	Those with severe knee OA had greater trunk lean (5 degrees) compared to those with no knee pain (1.6 degrees).
Birmingham et al. (2009)	128(102/26)	Patients with OA	47.5±9.5	Trunk lean decreased on average from 3.5 to 2.0 degrees pre and two years post high tibial osteotomy surgery.
Hunt et al. (2008)	120 (60/60)	Patients with OA	45.5±0.8	Trunk lean explained 13 of variance in first peak knee adduction moment.
Mundermann et al. (2008)	19(12/7)	Healthy	22.8±3.1	Walking with 10±5 degrees trunk lean reduced peak knee adduction moment by 65% compared to normal walking.
Tanaka et al. (2008)	12(0/12)	6 Unilateral OA 6 Bilateral OA	68.7±8.1	Patients with unilateral knee OA trunk leaned 3.7 degrees compared to 3.8 degrees in patients with bilateral knee OA and 4.4 and 3.9 degrees in control subjects.
Andriacchi & Mundermann (2006)				Trunk lean and toe-out are gait adaptations that could be ‘compensatory’ mechanisms to reduce loading in the weight bearing knee.
Mundermann et al. (2005)	84(38/46)	19 less severe OA 23 more sever OA 42 healthy control	65.2±12.5 65.0±8.0 61.7±12.3	Patients with OA are more likely to initiate a greater trunk lean through hip adductor muscle forces.

Table 1.2: Summary of findings from articles investigating toe-out

Authors	N (M/F)	Participants	Age	Primary Findings
Birmingham et al. (2009)	128(102/26)	Patients with OA	47.5±9.5	Toe-out increased from 12.0 to 13.2degrees pre and two years post high tibial osteotomy surgery.
Fregly et al. (2008)	1(1/0)	Patient with OA	41	Increasing toe-out by 15 degrees reduced second peak knee adduction moment by 38%.
Hunt et al. (2008)	120 (60/60)	Patients with OA	45.5±0.8	Toe-out angle explained 12% of variance in first peak knee adduction moment.
Jenkyn et al. (2008)	180(141/39)	Medial compartment OA	48.1(21-76)	Toe-out results in significant reductions in first and second peak knee adduction moment (11.7 and 34.4% respectively).
Lynn et al. (2008)	11(6/5)	Healthy	22.9±1.8	Increasing toe-out angle from 19 to 40 degrees resulted in a 12% increase and a 93% decrease in first and second peak knee adduction moment. Ten degrees toe in resulted in 64% increase in second peak knee adduction moment.
Lynn et al. (2008)	24(12/12)	12 Healthy 12 Knee OA	68.7±8.4 67.4±10	Exaggerated toe-out by 9.6 and 11 degrees resulted in reductions in second peak knee adduction moment in both healthy adults (22.5%) and patients with knee OA (42%) respectively.
Reinbolt et al. (2008)	1(1/0)	Medial compartment OA	41	Increased toe-out resulted in a 6% increase and 31% decrease in first and second peak knee adduction moment respectively.
Rutherford et al. (2008)	140(72/68)	50 Healthy 46 Mild to Moderate 44 Severe	53±10 60±9 67±8	Toe-out was associated with the knee adduction moment in only healthy and mild groups at second peak. Second peak is only seen in 30% of patients with severe and 60% of patients

				with mild OA.
Schache et al. (2008)	1(1/0)	Healthy	26	Increasing toe-out by 11 degrees reduced the knee adduction moment by 23% during second peak.
Chang et al. (2007)	56(23/33)	Patients with OA	66.6	Greater toe-out was associated with a lower likelihood of knee OA progression (Odds Ratio = 0.60).
Guo et al. (2007)	10(6/4)	Patients with OA	64±8	A 16.6 degree increase in toe-out translated to a 40% reduction in second peak knee adduction moment.
Hurwitz et al. (2002)	111(56/55)	62 with OA 49 healthy	62±10 59±10	Toe-out was significantly correlated with second peak knee adduction moment (r=-0.452).
Lin et al. (2001)	44	Healthy children	11-13	Increased toe-out from 10 to 20 degrees resulted in a 55% reduction in first peak knee adduction moment and 700% increase in second peak.
Andrews et al. (1996)	11(5/6)	Healthy	23-42	Higher toe-out angle reduced the second peak knee adduction moment (r=0.44).

1.6.2 Walking poles and Canes

Assistive walking devices such as canes are a common technique for individuals with disability to maintain independence, improve function, enhance safety, and protect joints (Van der Esch, Heijmans, & Dekker, 2003).

For patients with knee OA, the purpose of a cane, is in part to reduce knee joint loading and symptoms related to OA. Patients with symptomatic knee OA who use a cane, will carry the device on the contralateral side of the affected limb. By pressing down on the device, a moment is generated about the knee in the frontal plane that acts to abduct the femur relative to the tibia. Through the frontal plane force applied by the user and the long lever arm, the moment provided by the cane resists the knee adduction moment of the stance limb (Figure 1.5). This moment has been identified to be an important factor in estimating the effectiveness of walking devices on reducing knee joint load (Gross & Hillstrom, 2009). Kemp, Crossley, Wrigley, Metcalf, and Hinman (2008) investigated the effects of contralateral cane use on knee joint loading in forty patients with knee OA. They determined a 10% reduction in the knee adduction moment compared to walking unassisted. Chan, Smith, Kirtley, and Tsang (2005) evaluated the effects of cane placement in 14 patients with knee OA. They determined a 7% reduction in knee adduction moment with contralateral cane placement compared to unaided gait. Despite the positive effects of cane use, significant drawbacks have been identified such as reduced walking speed (Chan et al., 2005).

Nordic walking poles may act similarly to canes as users are encouraged to apply a load through the contralateral pole to the stance limb. Walking poles have recently been suggested as an option for individuals with disability to maintain a reasonable level of physical activity (Fregly et al., 2009; Oakley et al., 2008). Manufacturers have since incorporated populations with disability into marketing audiences as they are promoted to improve fitness while reducing stress on the joints of the lower limb (Urban Poling Inc., 2011). Walking poles however, could offer more benefit to patients with knee OA than other assistive walking devices. Walking with poles has been shown to decrease ground reaction force acting on the stance limb, increase walking speed, stride length, and cadence (Willson, Torry, Decker, Kernozek, & Steadman, 2001) while achieving a higher workload as measured by VO2 depending on surface type in healthy adults (Schiffer, Knicker, Dannohl, & Struder, 2009; Hansen & Smith, 2009). Investigations focusing on individuals with disability also report benefits. Oakley et al. (2008) investigated the effects of walking with poles on 20 patients with Intermittent Claudication, a circulatory disease of the lower extremities. They found patients walked further, with less pain, and higher workload without an increase in perceived exertion compared to walking without poles.

Findings from studies evaluating the effects of walking poles on knee joint loading vary substantially. Fregly, D'Lima, and Colwell (2009) investigated medial compartment direct contact force in one patient with an instrumented total knee replacement. They observed a 27% decrease in medial compartment contact force late in stance in one individual with an instrumented total knee replacement. Walter, D'Lima, Colwell, and Fregly (2010) also evaluated knee joint loading in one individual with an instrumented total knee replacement. They found a 33% and 47% decrease for first and second peak knee adduction moment respectively compared to walking without poles.

Conversely, Stief et al. (2008) investigated walking with poles on the lower extremities during gait in 15 healthy adults. They found a 15% increase in first peak knee adduction moment compared to unaided gait. Jensen et al. (2010) evaluated knee joint loading with different magnitudes of pole force in 10 healthy adults. They found no change in either first or second peak knee adduction moment between self-selected pole force and when force applied to the pole was increased by 2.4 times.

In summary, the benefits associated with walking poles may make them desirable to those dealing with the symptoms associated with knee OA. Their potential however, to reduce loading on the knee joint in patients with knee OA is currently unclear.

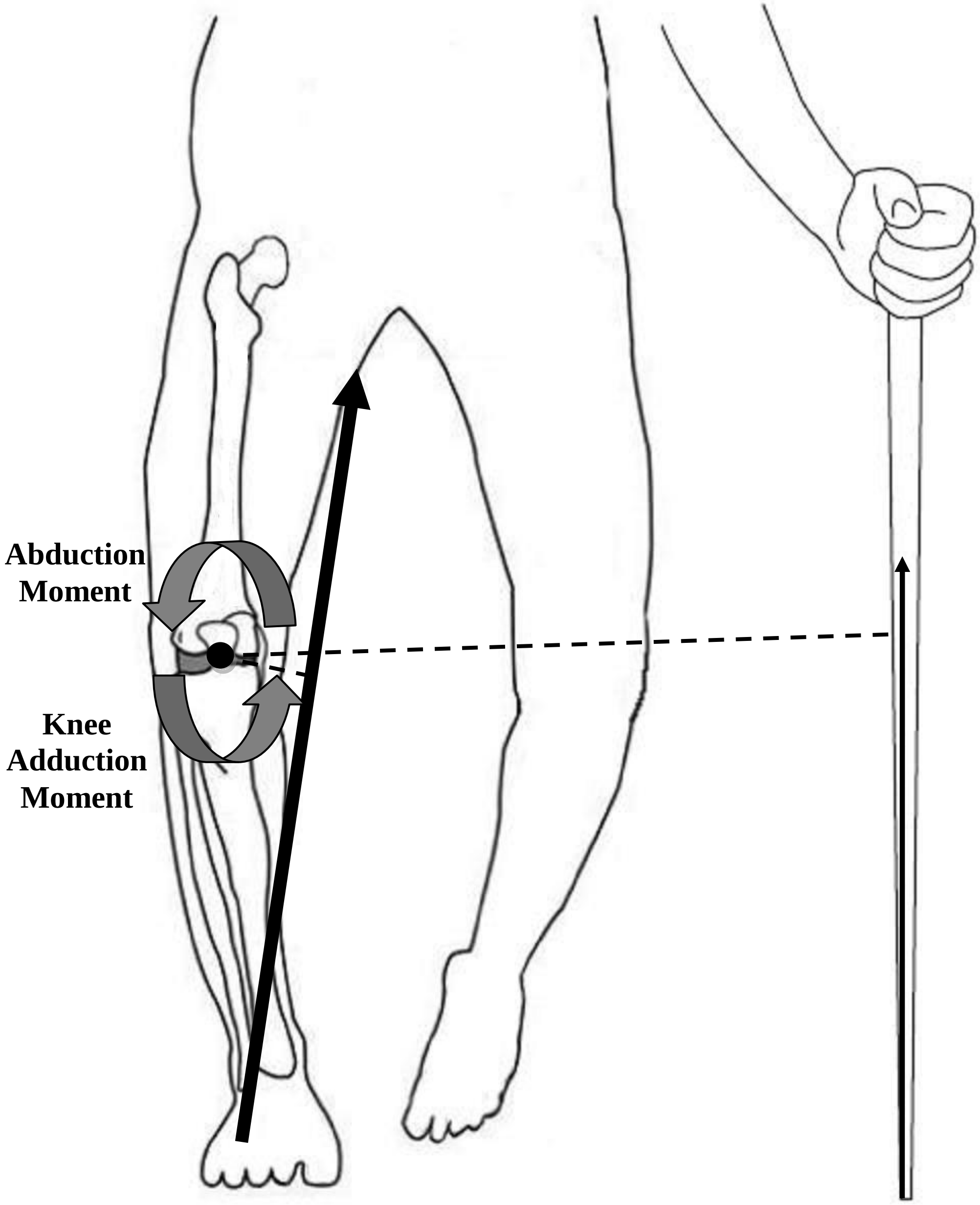

Figure 1.5. An illustration of the knee adduction moment and the moment generated by the walking pole during gait.

Table 1.3: Summary of findings from studies investigating walking poles and canes

Authors	N (M/F)	Participants	Age	Primary Findings
Jensen et al. (2010)	10(4/6)	Healthy	30	Knee adduction moment was unchanged when increasing pole force 2.4 times compared to self selected force
Walter et al. (2010)	1(1/0)	Patient with total knee replacement	83	First and second peak knee adduction moment decreased 33.1 and 47.2% respectively compared to unaided gait.
Fregly et al. (2009)	1/0	Patient with total knee replacement	83	Medial knee contact force was reduced by 27% in late stance when compared to walking without poles.
Schiffer et al. (2009)	13 (0/13)	Healthy	26±4	Peak axial pole force applied to the pole was 36.5 to 43.3 N depending on surface type.
Hansen and Smith (2009)	12 (1/11)	Healthy	50.6±2.4	Energy expenditure as high as 67% greater than walking without poles and 3% greater with 7.5cm shorter poles then self selected length.
Gross and Hillstrom (2009)	NA	NA	NA	Studies regarding knee joint loading should also consider the moment created by the assistive walking device that resists the knee adduction moment.
Hansen et al. (2008)	7 (0/7)	Healthy	51(42-58)	Walking with poles required as much as 67% greater energy expenditure than walking without poles.
Kemp et al. (2008)	40 (16/24)	Patients with OA	64.7±9.4	Cane associated with 10% reduction in knee adduction moment.
Oakley et al. (2008)	21 (21/0)	Patients with Intermittent Claudication	70(57-79)	Patients walked further, with less pain, at a higher work load, and no increase in perceived exertion.
Stief et al. (2008)	15 (15/0)	Healthy	31±4.6	Walking with poles resulted in 14% higher first peak and 2% lower second peak knee adduction moment

Bohne et al. (2007)	15 (15/0)	Healthy	(20-49)	Walking poles reduced sagittal plane moments for the ankle, knee, and hip while walking on a decline by 16.4, 10.6, and 9.7% respectively.
Chan et al. (2005)	14(0/14)	Patients with OA	NA	The use of canes caused slower walking speeds, and greater knee adduction moment when held on the ipsilateral side (40%) compared to an 8% decrease on the contralateral side.
Willson et al. (2000)	13 (8/5)	Healthy	29.5±5.1	Using walking poles increased gait speed, stride length, and cadence while decreasing ground reaction force compared to walking without poles.

1.7 Thesis Outline

The overall objective of this thesis was to explore the effects of prolonged walking and the use of Nordic walking poles on selected gait characteristics indicative of knee joint load. The thesis consists of a series of four studies. Data collection methods included both over-ground and treadmill three-dimensional gait analysis. Healthy participants had no disability or disease related to mobility, and were recruited from the surrounding community. Patients with knee OA were assessed and diagnosed by orthopaedic surgeons at the Fowler Kennedy Sport Medicine Clinic at the University of Western Ontario. Patients were recruited from the pool of potential participants being screened for an ongoing study investigating the effects of medial opening wedge high tibial osteotomy surgery. Individuals with knee OA were referred to this clinic mostly due to prolonged periods of unresolved pain localized to the knee. Data collected from healthy adults in study one (Chapter 2) were shared with study two (Chapter 3).

The objectives of study one and two were to investigate the effects of prolonged walking on gait patterns (selected kinematics) previously suggested to reduce knee joint load. The first study (Chapter 2) evaluated the time-varying behaviour, reliability, and validity of lateral trunk lean and toe-out angles during prolonged (60 min) treadmill walking in healthy adults. Study 2 (Chapter 3) used the methodology evaluated in Study 1 to compare the time-varying behaviour of lateral trunk lean and toe-out during prolonged (30 min) treadmill walking in healthy adults and patients with knee OA, and possible correlations with pain intensity. Treadmill time (i.e. the duration of walking) was reduced for patients with knee OA to mimic a realistic length of activity for patients with painful mobility.

The objectives of the third and fourth studies were to analyze the effects of a walking device (Nordic walking poles) on dynamic knee joint loading. The third study (Chapter 4) was a technical report that quantified frontal plane forces and moments generated by the pole about the knee in healthy adults. The fourth study (Chapter 5) used the methodology developed in the third study to evaluate the effects of walking poles on dynamic knee joint loading and related gait kinematics in patients with knee OA. The final chapter (Chapter 6) summarized the overall findings and offered direction for future investigations.

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Chapter 2:

Time-varying Behaviour, Test-retest Reliability and Concurrent Validity of Lateral Trunk Lean and Toe-out During Prolonged Treadmill Walking

2.1 Overview

Lateral trunk lean over the stance limb and toeing-out are potential adaptive gait mechanisms that reduce knee joint loading. The purpose of the present study was to evaluate the time-varying behaviour, the test-retest reliability and the concurrent validity of lateral trunk lean angle and toe-out angle during prolonged walking in healthy adults. Twenty healthy volunteers (51 ± 8 yrs, 12 females) completed two test sessions at least 24 hours apart but within the same week. For each participant, at each session, three-dimensional gait kinematics were assessed intermittently during 60 minutes of treadmill walking. Additionally, over-ground three-dimensional gait analysis was performed immediately before and after the treadmill walking. Maximum lateral trunk lean angle and maximum toe-out angle did not change over time ($p > 0.05$), were consistent from day to day (test-retest reliability: ICC = 0.61-to-0.72 and 0.90-to-0.95, respectively) and were consistent with over-ground measures (concurrent validity: ICC = 0.88 and 0.92, respectively). These findings suggest that lateral trunk lean angle and toe-out angle are consistent during prolonged walking and that these measures are reliable and valid for use in studying adaptive gait mechanisms.

2.2 Introduction

Several gait mechanisms identified using three-dimensional (3D) motion capture have been associated with decreased knee joint load during walking. In particular,

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increased lateral trunk lean over the stance limb (Andriacchi & Mundermann, 2006; Hunt et al., 2008; Hunt, Wrigley, Hinman, & Bennell, 2010; Mundermann, Dyrby, & Andriacchi, 2005; Mundermann, Asay, Mundermann, & Andriacchi, 2008; Tanaka et al., 2008) and increased toe-out (foot progression) angle (Andrews, Noyes, Hewett, & Andriacchi, 1996; Chang et al., 2007; Guo, Axe, & Manal, 2007; Hurwitz, Ryals, Case, Block, & Andriacchi, 2002; Jenkyn, Hunt, Jones, Giffin, & Birmingham, 2008; Rutherford, Hubley-Kozey, Deluzio, & Stanish, 2008) decrease the external adduction moment about the knee, which is a valid (Hurwitz, Sumner, Case, Block, & Andriacchi, 1998; Jackson et al., 2004; Thorp et al., 2006; Wada et al. 2001) and reliable (Andrews et al., 1996; Birmingham, Hunt, Jones, Jenkyn, & Giffin, 2008; Kadaba et al., 1989) proxy for load in the medial compartment of the tibiofemoral joint. Increasing the lateral lean of the trunk over the stance limb and increasing the toe-out angle of the stance foot both shift the ground reaction force (GRF) line of action closer to the centre of the stance knee. This serves to decrease the frontal plane lever arm of the GRF and reduce the knee adduction moment. Therefore, increased lateral trunk lean angle and toe-out angle have been both frequently discussed as gait mechanisms adopted by patients with knee osteoarthritis (OA) to lessen the load on the medial compartment of the joint (Andrews et al., 1996; Andriacchi, & Mundermann, 2006; Hunt et al., 2008; Mundermann et al., 2005; Mundermann et al., 2004; Tanaka et al., 2008; Hurwitz et al., 2002). Similarly, gait retraining to promote increased lateral trunk lean angle and toe-out angle have been discussed as possible therapeutic interventions for knee osteoarthritis (OA) (Chang et al., 2007; Mundermann et al., 2008; Guo et al., 2007).

Trunk lean has been consistently reported to be negatively associated with the knee adduction moment (typically first peak) in cohorts with and without knee OA (Andriacchi & Mundermann, 2006; Hunt et al., 2008; Mundermann et al., 2008; Mundermann et al., 2005; Tanaka et al., 2008). For example, when asked to walk with exaggerated lateral trunk lean, healthy participants walked with a mean of 10 degrees lateral trunk lean and reduced the knee adduction moment a mean of 1.29 %BW*Ht (approximately 65%) (Mundermann et al., 2008). Toeing-out during gait has also been consistently reported to be negatively associated with the knee adduction moment (typically second peak) in cohorts with (Chang et al., 2007; Guo et al., 2007; Hurwitz et al., 2002; Jenkyn et al., 2008; Rutherford et al., 2008) and without knee OA (Andrews et al., 1996; Teichtahl et al. 2006). For example, when participants were asked to walk with exaggerated toe-out (mean toe-out angle was 18.6 degrees) the knee adduction moment significantly decreased (mean decrease was 0.9% BW*Ht, approximately 38%) (Guo et al., 2007). Additionally, a smaller toe-out angle (≤4.4 degrees) in those with knee OA was associated with a greater likelihood of OA progression after 18 months (Chang et al., 2007).

Although lateral trunk lean and toe-out clearly affect knee joint loading, and likely influence disease progression, the potential compensatory nature of these mechanisms are presently unclear. If lateral trunk lean and toe-out angles are indeed compensatory, it is plausible that they will change during prolonged walking. Studies to date have evaluated trunk lean and toe-out exclusively using over-ground 3D gait analyses that typically involved only a limited number of steps and short walking trials.

Studies evaluating gait during prolonged walking are sparse and typically focus on temporal-distance characteristics (Matsas, Taylor, & McBurney, 2000; Wass, Taylor, & Matsas, 2005). The measurement properties of gait kinematics assessed during prolonged walking are unknown. Specifically, we are unaware of any studies evaluating the behaviour, reliability, or validity of trunk lean and toe-out over prolonged walking. Therefore, the purpose of the present study was to evaluate the time-varying behaviour, the test-retest reliability and the concurrent validity of lateral trunk lean angle and toe-out angles during prolonged walking in healthy adults. Specific objectives were to (1) evaluate the consistency of trunk lean angle and toe-out angle over 60 minutes of treadmill walking, (2) quantify the level of agreement between two test sessions completed on different days (test-retest reliability), and (3) quantify the level of agreement between these gait biomechanics as measured by treadmill and over-ground motion capture (concurrent validity). We hypothesized that trunk lean and toe-out would not change significantly during walking ($p>0.05$) and that trunk lean and toe-out would have good test-retest reliability ($ICC\geq0.85$) and concurrent validity ($ICC\geq0.85$).

2.3 Methods

2.3.1 Study Design

Participants were tested on two separate sessions completed at least 24 hours apart, but within the same week. This time period enabled estimates of day-to-day measurement error while minimizing the chance that true changes in gait occurred between testing sessions. During each test session, 3D gait kinematics were assessed intermittently during 60 minutes of treadmill walking. Over-ground 3D gait analysis was completed immediately before and after (within 2 minutes) the treadmill walking session. Measures were collected bilaterally; however, only right foot strikes were used. As such, each participant contributed gait data from one limb to the overall analysis. The study protocol was approved by the institution's Research Ethics Board for Health Sciences involving human participants and all participants provided written informed consent. Inclusion criteria included healthy adults 35 to 65 years of age reporting no existing health conditions likely to affect mobility such as existing joint pain.

2.3.2 Participants

Twenty volunteers, 8 male and 12 female, with no previously diagnosed health or mobility concerns participated in the study. Statistical justification for sample size was based on two test sessions and the parameter estimation of an intra-class correlation coefficient (ICC) of 0.85 with a 95% confidence interval (95% CI) width of 0.25 (Bonett, 2002).

Table 2.1: Demographics of the study participants (*n*=20).

Variable	**Mean**	**SD**	**Min-Max**
Age (year)	51	8	37 - 62
Height (m)	1.74	0.06	1.63 - 1.83
Mass (kg)	78.3	12.4	53.2 - 106.3
BMI (kg/m^2)	25.9	3.2	20.1 - 32.3
Gait Speed (Day 1)	1.3	0.2	1.1 - 1.8
Gait Speed (Day 2)	1.4	0.2	1.1 - 1.9

2.3.3 Over-ground Gait Analysis

During each test session, participants first underwent an over-ground gait analysis using an eight-camera motion capture system (Eagle HiRes cameras, EvaRT 4.2 system, Motion Analysis Corp., Santa Rosa, CA, USA) synchronized with a floor-mounted force plate. A modified Helen Hayes 22 passive reflective marker set was used (Kadaba, Ramakrishnan, & Wootten, 1990). An initial static trial was used to determine body mass, relative marker orientations, and the locations of joint centers for the knee and ankle. The participants wore their own comfortable shoes, shorts and a t-shirt. Participants were instructed to wear the same shoes for each test session. They were instructed to walk along an 8m long walkway at their self-selected preferred walking pace while kinematic (sampled at 60 Hz) data were collected. Over-ground walking trials were repeated until three complete right foot strikes on the force plate were obtained.

2.3.4 Treadmill Gait Analysis

Participants completed 60 minutes of continuous walking on a treadmill equipped with two force plates (Gaitway model, Kistler Instrument Corp., Amherst, NY, USA) synchronized with a four-camera motion capture system (Hawk cameras, EvaRT 4.2 system, Motion Analysis Corp., Santa Rosa, CA, USA). Walking speed was determined by averaging the forward velocity of the sacral marker over the middle two strides during a random over ground trial. Treadmill speed was previously calibrated and matched for each participant's self selected over ground walking speed. To ensure familiarization with the treadmill, participants were allowed to walk on the treadmill for up to 5 minutes prior to data collection. Data were recorded for the first 15 seconds of each five minute interval of the entire 60 minute walking period (i.e. 13 envelopes of data per treadmill test).

2.3.5 Data Reduction

All kinematic data were filtered using a 4th order Butterworth filter at 6 Hz.Trunk lean angle and toe-out angle were calculated from the kinematic data as previously described (Hunt et al., 2008; Jenkyn et al., 2008). Trunk lean angle was defined as the angle between vertical and the line connecting the midpoint of the anterior superior iliac spines (ASISs) to the midpoint of the anterior tips of the acromion processes. Positive values indicated a trunk lean towards the stance limb and negative values indicated towards the swing limb. The toe-out angle was defined as the angle between the line connecting the centre of the ankle to the head of the second metatarsal, and the line representing forward progression of the body. Positive values indicated toeing-out and negative values toeing-in.

For over ground gait analysis, the stance phase waveforms from the first three complete right foot strike trials were selected. The maximum lateral trunk lean angle during stance phase was identified from each trial and averaged. Toe-out angle was calculated in the same manner. For treadmill walking, the stance phase waveforms from the first three right foot strikes within each 15 second envelope were used. The first 15 second envelope (at 0 minutes) was excluded to reduce the potential effect of familiarization. The remaining envelopes were averaged to create four time windows. Data collected at 5, 10 and 15 minutes were averaged and labelled window one. Window two consisted of averaged data collected at 20, 25, and 30 minutes. Data collected at 35, 40, and 45 minutes were averaged and labelled window three. Window four consisted of averaged data collected at 50, 55, and 60 minutes.

2.3.6 Statistical Analysis

For objective 1, we averaged mean data from sessions one and two. We evaluated differences in lateral trunk lean during the four time windows of prolonged walking with one-way repeated measures analysis of variance (ANOVA). We evaluated toe-out angle in the same manner. For objective 2, we calculated intraclass correlation coefficients (ICC type 2,1) (Shrout & Fleiss, 1979) for each time window measured on separate days. We also calculated the standard error of measurement (SEM) (Steiner & Norman, 1995. For objective 3, we pooled data from sessions one and two and evaluated the agreement between over-ground and treadmill measures (concurrent validity) also using ICCs (type 2,1). Over-ground values collected before (pre) treadmill walking were compared to window one. Over-ground values collected after (post) treadmill walking were compared to window four.

2.4 Results

Participant demographics were reported in Table 2.1. The sample mean lateral trunk lean and toe-out angles with standard deviations were plotted in Figures 2.1 and 2.2, respectively over the entire 60 minutes of treadmill walking, in 5 minute intervals. The ANOVAs indicated no significant main effect for time for either of the lateral trunk lean ($F(3,57)=1.17$, $p=0.33$) and toe-out angles ($F(3,57)=2.84$, $p=0.05$). Although approaching statistical significance, the largest increase in toe-out angle was only 0.62° (Table 2.2). The average fluctuation (difference between maximum and minimum values during the 60 minutes session) in trunk lean angle was 1.3 ±0.5 degrees. The average fluctuation in toe-out angle was only 3.5±1.3 degrees. ICCs for trunk lean angle ranged from 0.61 to 0.75 (Table 2.2) and can be described as demonstrating moderate to good test-retest reliability (Portney & Watkins, 2000). ICCs for toe-out angle ranged from 0.90 to 0.95 (Table 2.2) and can be described as good reliability (Portney & Watkins). For trunk lean, 95% CI around the SEM ranged from 1.13 to 1.80 degrees. For toe-out, 95% CI around the SEM ranged from 2.18 to 3.01 degrees (Table 2.2). There was also good agreement (Portney & Watkins) between the over-ground and treadmill measures for trunk lean and toe-out angles with ICCs ranging from 0.88 to 0.92 respectively (Table 2.3).

Table 2.2: Agreement between testing days and standard error of measurement for lateral trunk lean and toe-out measures (in degrees) during prolonged treadmill walking divided into four time intervals.

Time Window	**Mean (SD) Test**	**Mean (SD) Retest**	**Mean Difference (95%CI)**	**ICC**	**SEM**
		Trunk lean angle (deg)			
1	0.66 (1.09)	0.77 (1.09)	-0.11 (-0.49, 0.27)	0.72	0.58
2	0.72 (1.20)	1.01 (1.42)	-0.30 (-0.78, 0.19)	0.71	0.65
3	0.81 (1.25)	0.94 (1.38)	-0.13 (-0.56, 0.30)	0.75	0.63
4	1.03 (1.48)	0.62 (1.27)	0.41 (-0.17, 0.98)	0.61	0.92
		Toe-out angle (deg)			
1	10.10 (4.84)	10.52 (4.96)	-0.42 (-1.45, 0.60)	0.90	1.53
2	10.59 (5.16)	10.69 (4.74)	-0.10 (-1.10, 0.90)	0.91	1.54
3	10.54 (4.98)	10.93 (5.29)	-0.39 (-1.17, 0.38)	0.95	1.11
4	10.72 (5.39)	10.90 (5.50)	-0.17 (-1.18, 0.83)	0.93	1.43

Table 2.3: Agreement between treadmill and over-ground walking measures. Trunk lean and toe-out (in degrees) values obtained during the first over-ground walking session (pre) are compared to window one of treadmill walking. Values obtained during window four of treadmill walking are compared to the second over-ground walking session (post).

Time	**Mean (SD)**		**Mean Difference (95% CI)**	**ICC**
	Treadmill	**Over-ground**		
		Trunk lean angle (deg)		
Pre	0.71 (1.01)	1.52 (1.01)	-0.80 (-1.04, -0.57)	0.88
Post	0.82 (1.24)	1.23 (1.08)	-0.41 (-0.67, -0.14)	0.88
		Toe-out angle (deg)		
Pre	10.31 (4.78)	9.52 (5.03)	0.79 (-0.15, 1.74)	0.92
Post	10.82 (5.27)	9.65 (5.08)	1.17 (0.17, 2.17)	0.92

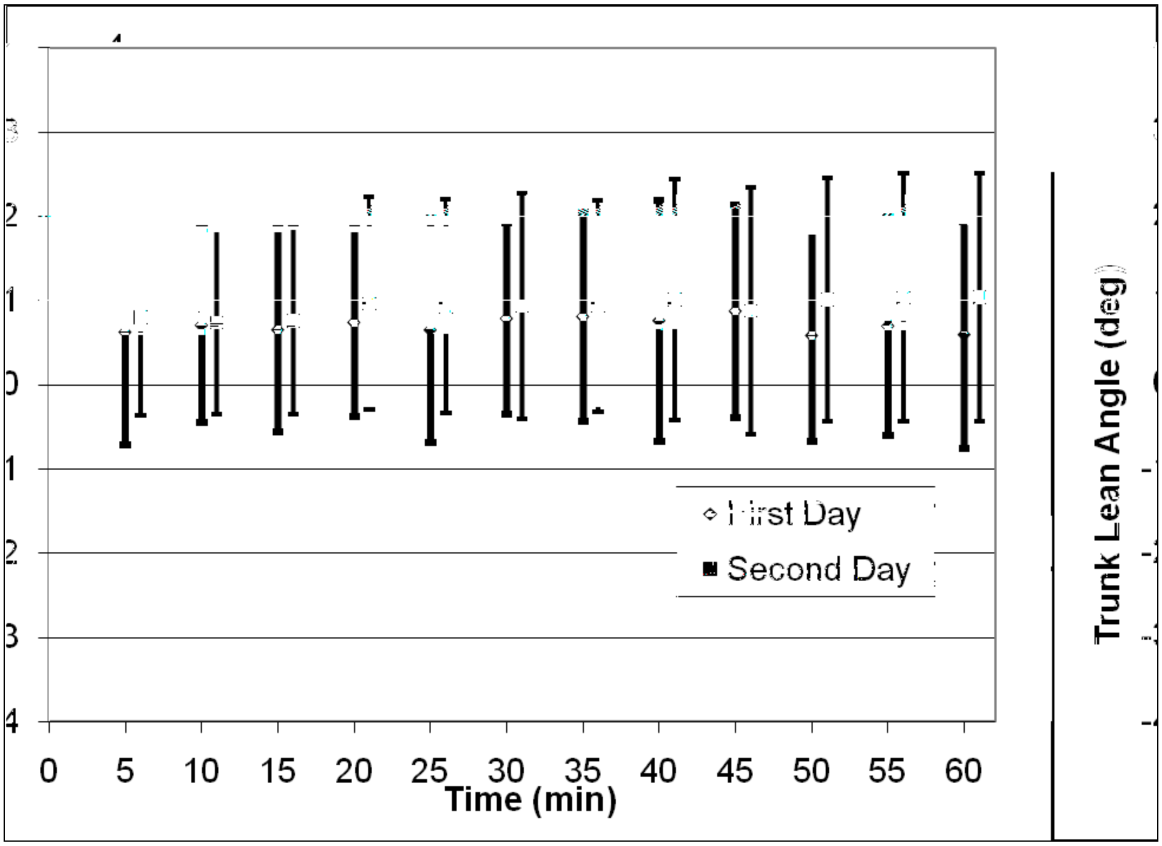

Figure 2.1. Mean lateral trunk lean angle with standard deviation bars plotted over the entire 60 minutes of treadmill walking in 5 minute intervals for both the first and second test sessions.

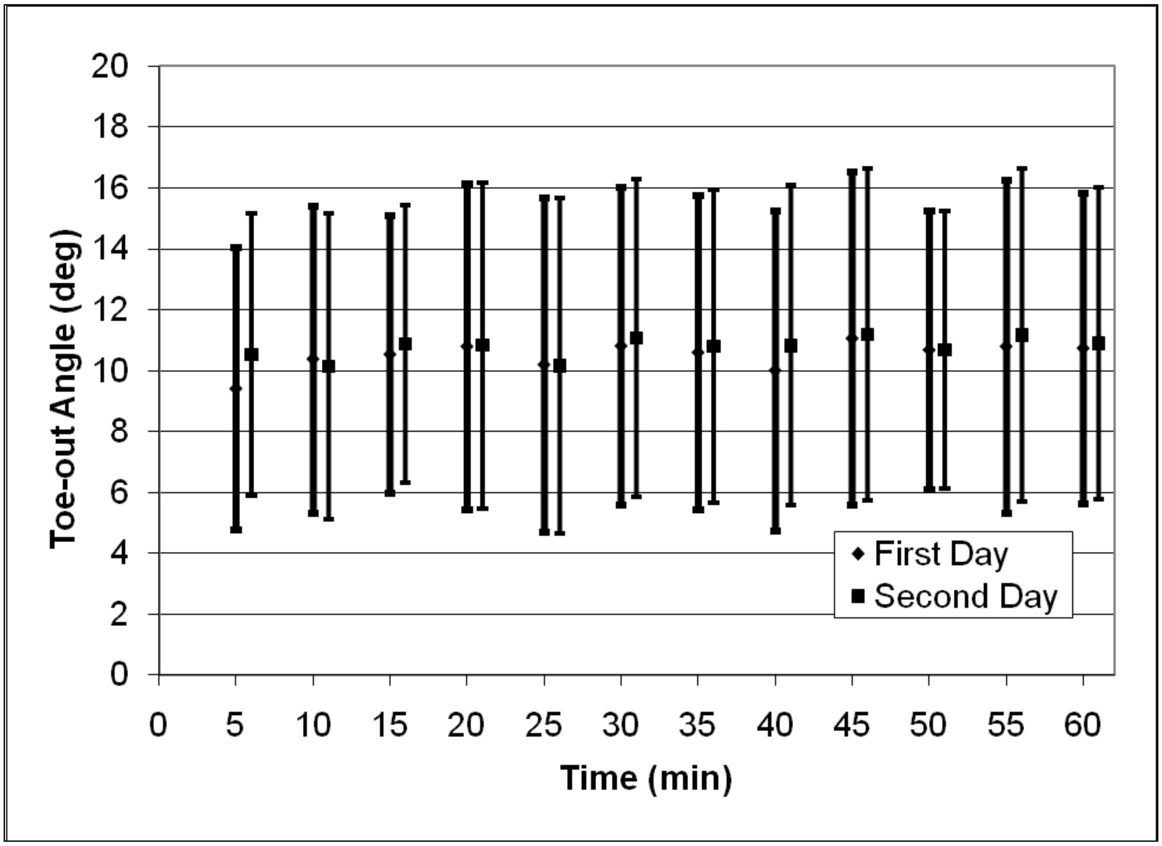

Figure 2.2. Mean toe-out angle with standard deviation bars plotted over the 60 minutes of treadmill walking in 5 minute intervals for the first and second test sessions.

2.5 Discussion

The present results indicate that lateral trunk lean and toe-out angles were consistent during prolonged treadmill walking in healthy participants. Although there were no statistical differences between time windows, small fluctuations were observed within each individual's data over the 60 minutes.

These results also suggest that maximum lateral trunk lean angle and maximum toe-out angle observed during prolonged treadmill walking in healthy participants were reliable from day to day (Table 2.2). Only one other study (Hunt et al., 2008) has investigated reliability of trunk lean and toe-out angles. Although Hunt et al. evaluated patients with knee OA during over-ground walking, their findings were similar to those of this study on healthy participants. They reported ICC values of 0.91 and 0.69 for trunk lean angle and toe-out angle respectively in a subgroup of 15 patients. Although the present findings suggest lower reliability for trunk lean and higher for toe-out, both studies suggest moderate to good reliability (Portney & Watkins, 2000) for these measures.

The present SEM results enable the estimation of individual measurement error in trunk lean and toe-out angles at various levels of confidence. For example, using the reported SEM (Table 2.2), present results suggest we can be 95% confident that an individual with a measured maximum trunk lean value of 0.66 degrees has a true value between 1.79 and -0.47 degrees. Although the present ICC's suggest high test-retest reliability, the SEM emphasizes caution should be adopted when interpreting potential changes in trunk lean and toe-out in individuals.

The angles for trunk lean and toe-out during treadmill walking are in agreement with those measured during over-ground walking, which is the current gold standard in gait analysis (ICCs = 0.88 and 0.92 for trunk lean and toe-out angles, respectively). This suggests that the lateral trunk lean angle and toe-out angle as measured during prolonged treadmill walking are indeed valid. However, the substantial mean differences reported in Table 2.3 (particularly in trunk lean measures) should be noted. Although the ICC's suggest high agreement between treadmill and over ground measures, and support concurrent validity, these methods do not appear to provide values that could be used interchangeably.

The present trunk lean angle results were similar to those measured by Hunt et al. (2010), who investigated differences in proximal body segment kinematics and kinetics in healthy participants and individuals with knee OA. They reported sample mean trunk lean angle of 1.6±2.4 and 0.9±2.0 degrees in 20 healthy participants. Trunk lean angle was defined using markers over the supra-sternal notch, and the posterior processes of the T2 and T10 vertebrae. The angles in the present study are lower than those measured in healthy participants investigated by Tanaka et al. (2008). They reported trunk lean angle of 4.4±1.9 and 3.9±0.8 degrees for left and right legs respectively for five healthy adults with an average age of 72 years. Potential explanations for the different results may be that study's lower sample size or the difference in age of participants. Tanaka et al. defined trunk lean as the difference between a line connecting S1 and C7 vertebrae and the vertical. The present results are also smaller than those observed in patients with knee OA. Birmingham et al. (2009) reported average trunk lean amongst 120 patients with knee OA undergoing high tibial osteotomy to be 3.45 ± 2.97 degrees. Higher trunk lean values in patients with knee OA have also been reported by others (Hunt et al., 2010; Mundermann et al., 2005).

The present sample mean toe-out angle is similar to previous studies that have investigated healthy individuals during over-ground walking. Lin et al. (2001) reported a mean of 10.0 ± 3.0 degrees in a group of 48 healthy teenagers, while Rutherford et al.

(2008) reported a mean toe-out of 7.5 ± 5.0 degrees in a group of 50 healthy adults. Both studies calculated toe-out angle in a similar manner to the present study.

Although treadmill walking enables the quantification of gait during prolonged walking, limitations must be acknowledged. Treadmill walking is different from over-ground walking in a number of ways, such as the requirement for increased coordination and the difference in proprioception during gait (Matsas et al., 2000; Riley, Paolini, Della Croce, Paulo, & Kerrigan, 2007; Wass et al., 2005). These differences may be exacerbated depending on the level of experience and age of the individual, but can also be minimized through proper familiarization with the treadmill (Matsas et al., 2000; Riley et al., 2007). Results of the present study can only be generalized to individuals similar to the present participants. These participants had no known pathology. It is unclear whether trunk lean and toe-out can be measured as reliably during prolonged walking in individuals with musculoskeletal conditions. In addition, the absence of disease was determined by self-report only. It is possible that some participants possessed some degree of pathology, but had no symptoms.

This is the first study to evaluate the behaviour and measurement properties of trunk lean and toe-out during prolonged walking. The present results suggest that treadmill walking can provide a suitable method for measuring trunk lean and toe-out angles over prolonged periods of walking. These results also provide normative data for future studies investigating gait adaptations during prolonged walking.

2.6 Conclusions

Maximum lateral trunk lean angle and toe-out angle during prolonged treadmill walking are consistent over time in healthy participants, are consistent from day to day and are comparable with over-ground measures. These results suggest that lateral trunk lean and toe-out angles during treadmill walking are reliable and valid for use in studying potential compensatory mechanisms.

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Chapter 3:

Toe-out, Lateral Trunk Lean and Pelvic Obliquity During Prolonged Walking in Patients with Medial Compartment Knee Osteoarthritis and Healthy Controls

3.1 Overview

Toe-out angle and lateral trunk lean are proposed compensatory gait mechanisms adopted by patients with medial compartment knee osteoarthritis (OA). The objectives of this study were to 1) compare time-varying behaviour of maximum toe-out angle, lateral trunk lean and pelvic obliquity during prolonged walking in participants with and without medial compartment knee OA, and 2) explore correlations between gait characteristics and pain intensity. Twenty patients with medial compartment knee OA and 20 healthy adults, completed 30 minutes of treadmill walking. Maximum toe-out angle, lateral trunk lean, pelvic obliquity and pain intensity were measured at 5 minute intervals. Toe-out was significantly smaller ($p=0.04$) in patients with knee OA (6.7±2.5º) than controls (10.3±2.2º). Toe-out significantly changed over time ($p<0.001$), but not in a systematic way with no significant interaction between group and time. Trunk lean was significantly higher ($p=0.03$) in patients with knee OA (2.0±1.0º) than controls (0.7±0.5º). Trunk lean did not change over time with no interaction between group and time. There were no statistically significant findings for pelvic drop. Pelvic rise was higher ($p=0.01$) in patients with knee OA (2.8±0.9°) than healthy controls (1.2±0.8°), but did not change over time with no significant interaction. Trunk lean and contralateral pelvic drop were significantly correlated to pain ($r = 0.65$ and 0.47, respectively). Although toe-out and trunk lean are different between those with and without knee OA, and patients with

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greater pain have greater trunk lean and contralateral pelvic drop, these gait characteristics do not serve as acute compensatory mechanisms.

3.2 Introduction

Approximately 17% of people over the age of 45 have symptomatic knee osteoarthritis (OA) (Lawrence et al., 2008). It is a leading cause of disability and increases the risk of disability due to other medical conditions substantially (Badley, 2005; Kadam & Croft, 2007). Accordingly, several authors stress the need to better understand risk factors and interventions for disease progression (Badley, 2005; Cooper et al., 2000; Wolf & Pfledger, 2003). Quantitative gait analysis may provide valuable insight. Gait compensations can be described as the tendency of patients to compensate for localized pain and pathology by adopting probable automatic gait modifications (Johnson & Waugh, 1979). Gait patterns that alter knee joint loads or are related to pain have recently engendered particular interest, especially toe-out angle (Andrew, Noyes, Hewett, & Andriacchi, 1996; Chang et al., 2007; Guo, Axe, & Manal, 2007; Hurwitz, Ryals, Case, Block, & Andriacchi, 2002; Jenkyn, Hunt, Jones, Giffin, & Birmingham, 2008; Lynn & Costigan, 2008; Rutherford, Hubley-Kozey, Deluzio, & Stanish, 2008), trunk lean towards the stance limb (Andriacchi & Mundermann, 2006; Hunt et al., 2008; Hunt, Wrigley, Hinman, & Bennell, 2010; Mundermann, Asay, Mundermann, & Andriacchi, 2008; Mundermann, Dyrby, & Andriacchi, 2005; Tanaka et al., 2008) and associated pelvic obliquity (Linley, Sled, Culham, & Deluzio, 2010). These gait patterns are correlated with the external adduction moment about the knee (Andrews et al., 1996; Chang et al., 2007; Guo et al., 2007; Hurwitz et al., 2002; Jenkyn et al., 2008; Lynn &

Costigan, 2008; Rutherford et al., 2008; Andriacchi & Mundermann, 2006; Hunt et al., 2008; Hunt et al., 2010; Mundermann et al., 2008; Mundermann et al., 2005; Tanaka et al., 2008).

Some authors suggest that these gait mechanisms are adopted by patients with knee OA to unload the joint and lessen symptoms. Wang et al.(1990) propose that toe-out is an adaptive mechanism used by some patients with varus gonarthrosis, suggesting that this adaptive mechanism reduces after surgical intervention such as a high tibial osteotomy. Chang et al.(2007) suggest that patients with OA who walk with greater toe-out have decreased likelihood of radiographic disease progression. We are aware of only one study comparing toe-out in those with and without knee OA. Lynn et al. (2008) report no difference in maximum toe-out angle during stance between patients with knee OA and healthy controls.

When discussing secondary gait modifications in patients with knee OA, Mundermann et al. (2005) suggest that shifting the trunk over the stance limb is also a possible gait compensation in those with knee OA, not seen in those without OA. Briem et al. (2009) also suggest proximal adaptations in those with knee OA. Hunt et al. (2010) support this, reporting smaller trunk lean angle in healthy adults compared to individuals with mild and moderate knee OA. Linley et al. (2010) also report differences between patients with knee OA and healthy adults, although these differences were only apparent when using advanced statistical techniques. We are aware of only one study reporting the relationship between toe-out and trunk lean and pain. Hunt et al. (2008) found a weak correlation between maximum toe-out angle and WOMAC pain score, and a low but significant correlation between maximum lateral trunk lean and WOMAC pain score. The role of pelvic obliquity in patients with knee OA has also gained recent attention. Although not argued to be a direct compensatory mechanism, pelvic obliquity has been discussed as a contributor to lateral trunk lean (Hunt et al., 2008; Hunt et al., 2010; Linley et al., 2010; Mundermann et al., 2008).

As described, previous research clearly suggests that toe-out, lateral trunk lean, pelvic obliquity and pain affect knee joint loading. Differences between individuals with and without knee OA, and the relationship between these gait patterns and pain, are less clear. Importantly, the changes of these gait patterns over prolonged walking is currently unclear. If these gait patterns are indeed compensatory, it is plausible that changes may be observed during prolonged walking in individuals with knee OA that are not observed in healthy adults. The purposes of this study were to 1) compare the time-varying behaviour of toe-out angle, lateral trunk lean and pelvic obliquity during 30 minutes of treadmill walking in participants with and without knee OA, and 2) explore correlations between these gait characteristics and perceived levels of pain. We hypothesized that toe-out would be lower in patients than controls and would increase over time in patients only. Similarly, we hypothesized that trunk lean and pelvic obliquity would be greater in those with knee OA and would increase over time in patients only. In other words, we hypothesize that the effect of time on toe-out and trunk lean would depend on group. We also hypothesized that all variables would be significantly correlated with pain intensity for patients with knee OA.

3.3 Methods

3.3.1 Participants

Twenty healthy adults (8 male) and 20 patients with knee OA (12 male) volunteered for this study. Patients were recruited from a tertiary care clinic specializing in orthopaedics. Patients were seeking medical consultation for unresolved knee pain localized in the medial compartment of the knee. All patients were diagnosed with varus gonarthrosis, defined as having varus alignment and OA confined mainly to the medial compartment. Using hip to ankle, frontal plane x ray, diagnosis was based on the Altman criteria (Altman et al., 1986) and radiographic severity was graded on the Kellgren and Lawrence (KL) scale (Kellgren & Lawrence, 1957). Exclusion criteria included inflammatory or infectious arthritis of the knee, neurological or psychological condition, unable to speak/read English, or inability to understand informed consent. Healthy adults were recruited from the surrounding community through print advertising and word of mouth. Exclusion criteria included any pre-existing health conditions likely to affect mobility, unable to speak/read English, or unable to understand informed consent. Sample size was based on the ability to detect a significant interaction ($p<0.05$) between group (patients with knee OA and healthy controls) and time (5 to 30 minutes) of medium effect size ($f = 0.2$), 95% of the time (G*Power Version 3.1.1, Universitat Kiel, Germany). This study was approved by the institution's Research Ethics Board for Health Sciences involving human participants. All participants provided written informed consent.

Table 3.1: Demographics of the study participants (*n*=40), listing mean (standard deviation).

	Healthy	Patients	Overall
Age (years)	51 (8) [37, 62]	55 (8) [44, 72]	53 (8) [37, 72]
Height (m)	1.74 (0.06) [1.63, 1.83]	1.74 (0.10) [1.50, 1.88]	1.74 (0.80) [1.50, 1.88]
Mass (kg)	78.3 (12.4) [53.2, 106.3]	87.2 (11.8) [63.7, 106.5]	82.8 (12.8) [53.2, 106.5]
BMI (kg/m^2)	25.9 (3.2) [20.1, 32.3]	28.9 (3.0) [24.1, 34.0]	27.4 (3.1) [20.1, 34.0]
Dynamic lower limb alignment (°)*	-0.86 (3.78) [5.83, -7.51]	-6.05 (3.76) [3.76, -11.09]	
Mechanical Axis Angle (°)*	-	-6.1 (3.2) [-0.7, -12.9]	
Males	8	12	20
Females	12	8	20
Kellgren and Lawrence grade, number of patients			
2		9	
3		10	
4		1	

BMI = Body Mass Index
***Negative values indicate varus alignment**

3.3.2 Gait Analysis

All participants were tested on one occasion in which three-dimensional (3D) gait kinematics were intermittently sampled (60Hz) during 30 minutes of treadmill walking. The participants wore comfortable shoes, shorts and a t-shirt. We used a modified Helen Hayes marker set of 22 passive reflective markers (Kadaba et al., 1989). Additional markers were placed over the medial knee joint line and medial malleolus bilaterally while participants stood on the force plate to determine body mass, marker orientation and positions of joint centers of rotation for the knee and ankle. The additional markers were removed prior to gait testing. Participants were instructed to walk along an 8m long walkway at their self-selected preferred walking pace to determine gait speed (based on the sacral marker) to be used during treadmill walking. Dynamic lower limb alignment was determined from the over-ground gait analysis as previously described (Hunt, Birmingham, Jenkyn, Giffin & Jones, 2008). All participants then completed 30 minutes of continuous walking on a force plate instrumented treadmill, (Gaitway model, Kistler Instrument Corp., Amherst, NY, USA) synchronized with a six-camera, high resolution motion analysis system (Hawk cameras, EvaRT 5.0 system, Motion Analysis Corp., Santa Rosa, CA, USA). To ensure familiarization, participants walked on the treadmill for a minimum of 5 minutes prior to testing. After a brief rest period, testing commenced when the treadmill speed matched the over-ground walking speed. Data were recorded for the first 15 seconds of each 5 minute interval from time 0 to 30 minutes. Pain intensity was quantified at these same intervals using an eleven point numerical rating scale (NRS). Participants were asked to rate the pain in their affected knee with 0 indicating no pain and 10 representing the worst pain imaginable. In the instance of bilateral knee OA, the limb of interest was the knee that was most symptomatic. In healthy adults, analysis was confined to the right limb.

3.3.3 Data Reduction

Maximum toe-out angle was calculated as the angle between a line drawn between the centre of the ankle and the head of the second metatarsal and the line of forward progression of the body. Positive values indicated toeing-out while negative values indicated toeing-in. The maximum lateral trunk lean angle was calculated as the angle of a line drawn from the midpoint of the anterior superior iliac spines (ASISs) to the midpoint of the anterior tips of the acromion processes with respect to the vertical. Positive values indicated a trunk lean towards the stance limb, while a negative value indicated a lean towards the swing limb. The maximum pelvic obliquity was calculated as the angle between the ASIS markers with respect to the horizontal. Positive values indicated a pelvic hike of the ASIS marker on the side of the swing limb while negative values represent a drop of the ASIS marker on the side of the swing limb. For each individual, maximum toe-out, lateral trunk lean and pelvic obliquity angles during stance phase were identified and averaged for the first three successive foot strikes for each 15 second interval. The first 15 second interval (time point 0) was excluded to reduce the risk of a familiarization effect. Data reduction and post processing were completed using commercially available (Orthotrak 4.2.1, Motion Analysis Corp., Santa Rosa, CA, U.S.A.) and custom software.

3.3.4 Statistical Analysis

We first calculated descriptive statistics (means, standard deviations, 95% confidence intervals) for participant demographics, pain and kinematic variables of interest. We evaluated pain intensity in patients with knee OA over time using one-factor repeated measures ANOVA. We compared toe-out, trunk lean and pelvic obliquity between groups (knee OA vs. healthy controls) and over time (5, 10, 15, 20, 25, 30 minutes) using two-factor repeated measures ANOVAs. In the event of a significant main effect, Scheffe post hoc tests were planned. We examined the relationships between toe-out, trunk lean, pelvic hike, pelvic drop and pain intensity at 5 minutes and 30 minutes of walking using Pearson correlation coefficients. Data were reported using sample averages and 95% confidence intervals.

In exploratory, post hoc analyses, we repeated the ANOVAs using BMI as a covariate. We then repeated the analysis using dynamic lower limb alignment as a covariate. Finally, we explored the differences between severity groups by repeating the ANOVAs while replacing the patient vs. control between-groups factor with a more-severe (KL grade 3 or 4) vs. less-severe (KL grade 2) between-groups factor. Statistical analyses were performed using commercially available software (Statistica 7.0, StatSoft Inc., Tulsa, OK, U.S.A).

3.4 Results

Participant demographics are reported in Table 3.1. Descriptive statistics for each group and their differences for maximum toe-out angle, maximum lateral trunk lean, maximum pelvic rise and drop and pain at each time interval are reported in Table 3.2. Ensemble averages of all time intervals for all participants in both groups are illustrated in Figure 3.1. Time-varying sample averages for all kinematic variables are reported in Figure 3.2. Time-varying sample averages for pain intensity are reported in Figure 3. All participants in the control group rated pain as 0 throughout testing. For patients with knee OA, ANOVA indicated a main effect for pain ($F(6, 114)=6.48$, $p<0.001$). Post hoc analysis showed pain at 0 minutes was significantly lower than pain at 10, 15, 20, 25 and 30 minutes.

There was no significant interaction between group and time for toe-out ($F(5, 190)=0.63$, $p=0.68$). There were significant main effects for both group ($F(1, 38)=4.66$, p=0.04) and time (F(5, 190)=4.04, p=0.002). Toe-out angle was lower in patients with knee OA (6.7±2.5 degrees) versus healthy adults (10.3±2.2 degrees) at all time points. Post-hoc tests showed toe-out angle at 5 minutes was significantly lower than at 15 (p=0.04) and 20 minutes (p=0.04). The average fluctuation within each individual's data over the 30 minute session (difference between highest and lowest maximum toe-out values during the 30 minutes session) was <3.5 degrees. Toe-out was not related to pain.

There was no significant interaction between group and time for lateral trunk lean (F(5, 190)=0.51, p=0.77). There was a significant main effect for group (F(1, 38)=5.40, p=0.03) but not for time (F(5, 190)=0.32, p=0.90). Lateral trunk lean was higher in patients with knee OA (2.0±1.0 degrees) compared to healthy adults (0.7±0.5 degrees) at all time points. Lateral trunk lean was related to pain intensity at both start and end time points (r = 0.65 at 5 min and r = 0.64 at 30 min).

There was no significant interaction between group and time for pelvic drop (F(5, 190)=0.17, p=0.97). There was no significant main effect for group (F(1, 38)=2.24, p=0.14) or time (F(5, 190)=0.35, p=0.74). Patients with knee OA tended to have less pelvic drop (-3.7±1.3 degrees) compared to healthy adults (-5.0±1.0 degrees) at all time points although differences did not reach statistical significance. Pelvic drop was related to pain intensity at 5 minutes (r = 0.47), but not at 30 minutes. There was no significant interaction between group and time for pelvic rise (F(5, 190)=1.34, p=0.25). There was a significant main effect for group (F(1, 38)=6.74, p=0.01) but not for time (F(5, 190)=0.50, p=0.78). Patients with knee OA had greater pelvic rise (2.8±0.9 degrees) than healthy adults (1.2±0.8 degrees) at all time points. Individual average fluctuations in trunk lean angle, pelvic drop and pelvic rise were all below 1.6 degrees. Pelvic rise was not related to pain intensity.

Results of the exploratory, post-hoc analyses suggested that the differences between patients and healthy controls for toe-out (p=0.04) and trunk lean (p=0.04) still existed when controlling for BMI, but not when controlling for dynamic lower limb alignment (p=0.09 and 0.26, respectively). Exploratory analyses did not detect differences in the behaviour of the observed gait characteristics in patients with KL grade 2 versus grades 3 or 4.

Table 3.2: Summary statistics, mean difference and 95% confidence intervals for toe-out, trunk lean and pelvic obliquity measures during 30 minutes of treadmill walking for both healthy adults and patients with knee osteoarthritis.

	Healthy Mean (SD)	**Patients Mean (SD)**	**Difference (95%CI)**
Toe-out angle (deg)			
5	9.4 (5.0)	6.2 (6.1)	3.2 (-0.4, 6.7)
10	10.4 (4.6)	6.5 (6.1)	3.9 (0.5, 7.3)
15	10.5 (5.4)	7.3 (5.9)	3.3 (-0.3, 6.9)
20	10.8 (5.5)	7.0 (5.9)	3.8 (0.2, 7.4)
25	10.2 (5.2)	6.2 (5.6)	4.0 (0.5, 7.4)
30	10.8 (5.2)	6.9 (5.8)	3.9 (0.4, 7.4)
Trunk lean angle (deg)			
5	0.6 (1.2)	2.0 (2.0)	-1.4 (-2.4, -0.3)
10	0.7 (1.2)	1.9 (2.1)	-1.2 (-2.3, -1.1)
15	0.7 (1.1)	2.0 (1.8)	-1.4 (-2.6, -0.1)
20	0.7 (1.3)	1.8 (2.2)	-1.1 (-2.2, 0.1)
25	0.6 (1.1)	2.0 (2.5)	-1.4 (-2.6, -0.1)
30	0.8 (1.2)	2.1 (2.2)	-1.3 (-2.4, -0.2)
Contralateral pelvic drop (deg)			
5	-4.9 (2.5)	-3.6 (2.4)	-1.3 (-2.8, 0.3)
10	-5.0 (2.3)	-3.9 (2.9)	-1.1 (-2.7, 0.6)
15	-5.0 (2.4)	-3.8 (3.2)	-1.2 (-3.0, 0.6)
20	-5.0 (2.5)	-3.7 (3.1)	-1.3 (-3.1, 0.5)
25	-5.0 (2.5)	-3.7 (3.3)	-1.4 (-3.2, 0.5)
30	-5.0 (2.3)	-3.7 (3.3)	-1.3 (-3.1, 0.5)
Contralateral pelvic rise (deg)			
5	1.4 (2.1)	2.6 (2.2)	-1.2 (-0.3, 0.2)
10	1.2 (2.0)	2.5 (2.2)	-1.3 (-2.6, 0.0)
15	1.2 (2.0)	2.9 (2.1)	-1.8 (-3.1, -0.5)
20	1.2 (2.0)	2.8 (2.1)	-1.6 (-2.9, -0.3)
25	1.1 (1.8)	2.9 (2.1)	-1.8 (-3.1, -0.5)
30	1.1 (1.8)	3.0 (2.0)	-1.9 (-3.1, -0.6)
NRS pain scale*			
0		1.6 (2.1)	
5		2.0 (2.1)	
10		2.3 (2.2)	
15		2.5 (2.1)	
20		2.5 (2.1)	
25		2.5 (2.0)	
30		2.5 (2.0)	

*** NRS = Numeric Rating Scale**

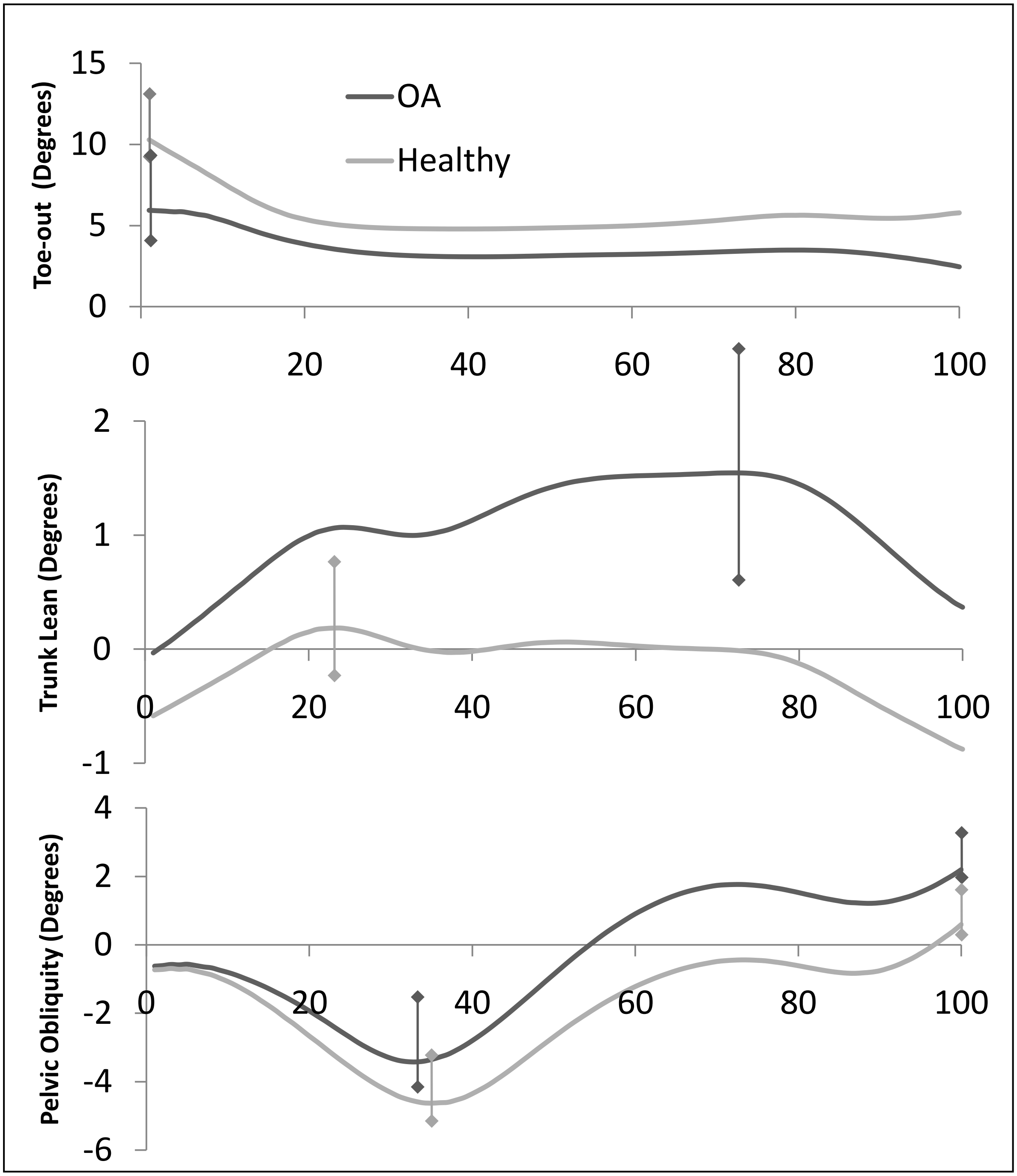
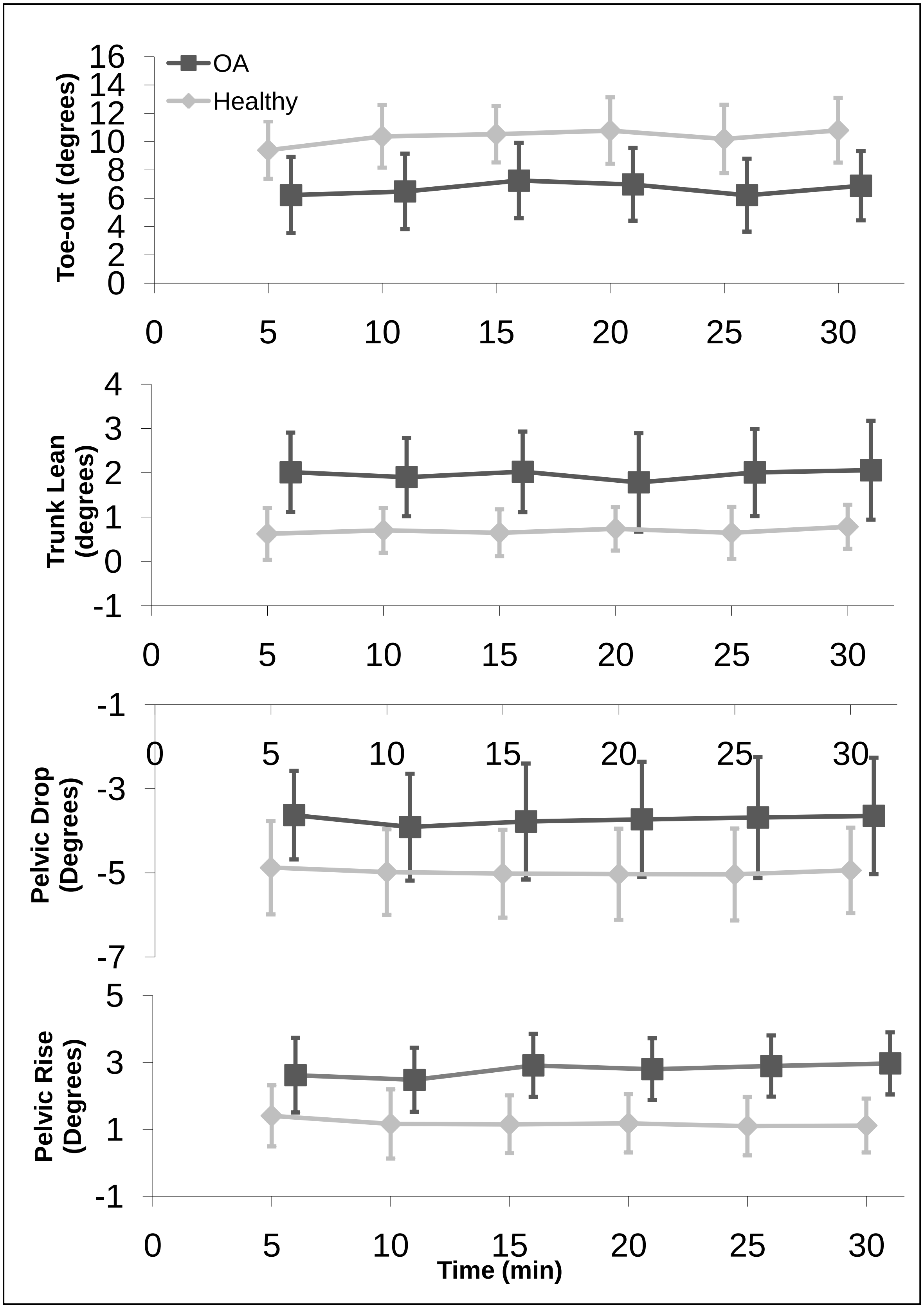


Figure 3.1. Ensemble averages for toe-out angle, lateral trunk lean angle and pelvic obliquity angle. Averages were calculated by normalizing each trial to 100% of stance for each in both patients with osteoarthritis and healthy adults. Ninety-five percent confidence intervals are indicated where data were sampled.



OA
Healthy
Toe-out (degrees)
16
14
12
10
8
6
4
2
0
0
5
10
15
20
25
30
Trunk Lean (degrees)
4
3
2
1
0
-1
0
5
10
15
20
25
30
-1
0
5
10
15
20
25
30
Pelvic Drop (Degrees)
-3
-5
-7
5
3
1
-1
Pelvic Rise (Degrees)
0
5
10
15
20
25
30
Time (min)

Figure 3.2. Time-varying behaviour of toe-out, trunk lean, pelvic drop and pelvic rise over 30 minutes of prolonged treadmill walking for groups with and without knee osteoarthritis. Data points include sample means at each time point +/- 95% confidence intervals.

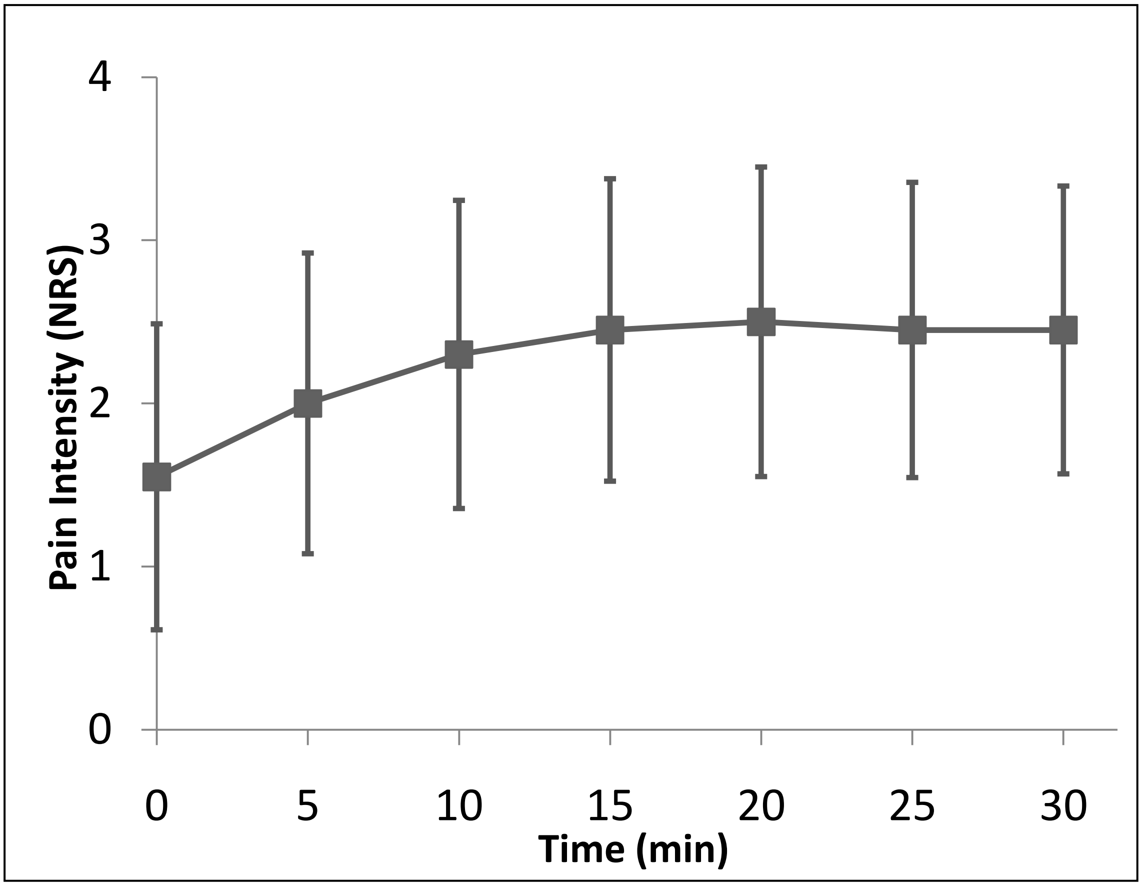


Figure 3.3. Time-varying behaviour of pain intensity of 30 minutes of prolonged treadmill walking for individuals with knee OA. All healthy participants reported no knee pain throughout testing. Data points include sample means at each time point +/- 95% confidence intervals.

3.5 Discussion

The primary objective of this study was to compare time-varying behaviour of toe-out angle, lateral trunk lean and pelvic obliquity during 30 minutes of prolonged walking in participants with and without knee OA. Our results do not support the hypothesized increases in toe-out, trunk lean and pelvic obliquity over time in patients with knee OA.

Although we did observe significant variation in mean toe-out angles at two time points, these changes were small and not systematic. The present results do suggest that patients with knee OA have significantly lower toe-out values than healthy adults. We are aware of only one previous study that compared toe-out in participants with and without knee OA (Lynn & Costigan, 2008). The present toe-out values (Table 3.2) are similar to those reported by Lynn and Costigan who investigated the role of toe-out on knee kinetics and hamstring activation. Although their results did not reach statistical significance, they also observed differences between 12 patients with knee OA (7.5±5.9 degrees) and 12 healthy adults (11.5±4.7 degrees). The higher number of participants and observations per individual in the present study provided greater power to detect statistical differences.

The present results suggest that lateral trunk lean and pelvic obliquity angle are consistent over prolonged treadmill walking. The results of the current study also suggest that patients with knee OA have significantly greater maximum lateral trunk lean and pelvic rise compared to healthy adults (Table 3.2). Although we also observed small differences in pelvic drop, these did not reach statistical significance. Overall patients have a greater pelvic obliquity angle throughout stance, but differences between groups are very small (<2 degrees). The clinical importance of such small differences is unknown.

The present results are in agreement with previous investigations that make direct comparisons in trunk lean between patients with knee OA and healthy adults. Those with knee OA adopt a greater magnitude of lateral trunk lean than their respective healthy cohorts. For example, Hunt et al.(2010) investigated differences in trunk lean in individuals with knee OA and healthy participants. Consistent with the present study, they reported sample mean trunk lean angle of 1.6±2.4 degrees in healthy adults compared to 2.3±2.0 and 3.1±2.0 degrees in individuals with mild and moderate knee OA respectively.

Several recent studies have also investigated the contribution of the pelvic segment to trunk lean (Hunt et al., 2008; Hunt et al., 2010; Linley et al., 2010). Mundermann et al. (2005) reported lower external hip adduction moments in patients with severe knee OA compared to less severe and healthy controls. The authors attributed this result to possible hip abductor muscle weakness in patients with severe knee OA that led to a pelvic tilt towards the swing limb. Although trunk and pelvic motion were not reported, the authors suggested that pelvic drop over the swing limb could lead to a trunk lean away from the stance limb resulting in higher knee joint loading. Hunt et al. (2010) found no significant differences in pelvic drop despite an increase in trunk lean over the stance limb between individuals with knee OA and healthy adults. Through retrospective analysis, they suggested that increasing trunk lean over the stance limb appeared to be associated with raising the hip marker over the swing limb. Those observations appear to agree with the current study as patients with knee OA walked with less pelvic drop and greater pelvic rise than healthy adults.

Importantly, the lack of systematic changes over time in toe-out, trunk lean and pelvic obliquity occurred despite a significant increase in pain intensity in patients with knee OA (Figure 3.3). The secondary objective of this study was to explore correlations between these gait characteristics and pain intensity. Our hypothesis was only confirmed for trunk lean and pelvic drop. In the only study we are aware of that investigated the relationship between toe-out, trunk lean and pain, Hunt et al. (2008) found a low but significant correlation (R=-0.18) between trunk lean and WOMAC pain subscale score. Toe-out measures did not result in the same relationship. Pelvic obliquity was reported, but not correlated to pain.

No change over 30 minutes of treadmill walking and significant correlation with pain intensity suggest lateral trunk lean is not adapted over time; although, patients who report greater pain have greater trunk lean. Toe-out angle and pelvic obliquity were not adapted through prolonged walking or related to pain. It may be that gait mechanisms such as increased trunk lean do not serve as an acute compensatory mechanism, but rather represent a walking mechanism that develops over long periods of time that parallel the symptoms of disease progression. Despite this possibility, few prospective observational studies exist that report the same gait variables as the present study. Chang et al. (2007) determined that a smaller toe-out angle in those with knee OA was associated with a greater likelihood of OA progression after 18 months. Miyazaki et al. (2002) found those with more severe pain and higher knee joint loading at baseline were more likely to show disease progression six years later. Neither study reported changes in toe-out or trunk lean. Birmingham et al. (2009) reported increased toe-out and decreased trunk lean two years post high tibial osteotomy that corresponded to a reduction in knee joint loading. Pelvic obliquity was not reported. It is unclear from this study what proportion of change in toe-out and trunk lean can be attributed to post-operative anatomical realignment and what proportion to factors such as change in pain.

It is possible that the differences between patients and controls in the gait characteristics we evaluated are related to the differences in body size and lower limb alignment in these groups. Our exploratory analyses suggest that differences in lower limb alignment are particularly important. Specifically, the observed between-group differences in trunk lean and toe-out no longer exist when controlling for the differences in lower limb alignment. This finding is consistent with recent reports that emphasize the importance of malalignment to increased compartment loading (Moyer, Birmingham, Chesworth, Kean & Giffin, 2010; Robbins, Birmingham, Maly, Chesworth & Giffin 2011) and to disease progression (Moisio et al., 2011). It is also possible that patients with greater disease severity exhibit different changes in gait during prolonged walking than patients with lesser severity. Although our exploratory analysis did not detect such differences, the limited sample size for this subgroup analysis (n=11 and 9 respectively) should be noted.

Over ground gait analysis is commonly used to measure gait kinematics and kinetics. However, data collected in that manner only represents a single moment in time. Since patients with knee OA commonly exhibit symptoms of activity onset pain, prolonged continuous treadmill walking could be argued to better mimic everyday activity. It must be acknowledged however, that despite its advantages, treadmill walking is different from over ground walking with respect to a number of different factors (Matsas, Taylor, & McBurney, 2000; Riley, Paolini, Della Croce, Paulo, & Kerrigan, 2007; Wass, Taylor, & Matsas, 2005). This is most likely the result of the higher level of coordination required and differences in proprioception. These differences can be mitigated through proper familiarization techniques (Matsas, Taylor, & McBurney, 2000;

Wass, Taylor, & Matsas, 2005). Regardless of these limitations, the treadmill is valid and reliable tool (Bechard, Birmingham, Zecevic, & Jenkyn, 2011) that provides a suitable method for measuring certain gait kinematics over prolonged periods of time.

Potential limitations in this study include the possibility that some healthy participants possessed some degree of knee OA, but had no symptoms. Also, patients with knee OA reported only mild to moderate pain and the mean increase in pain with prolonged walking was rather small. It is possible that results may be different in patients who experience greater (perhaps earlier) increases in pain during walking. As a result, the present study can only be generalized to individuals similar to the present participants.

This is the first study we are aware of that evaluated the behaviour of gait mechanisms during prolonged treadmill walking in patients with knee OA and healthy adults. Our results suggest that toe-out, trunk lean and pelvic obliquity are different between these groups, but are consistent over prolonged walking despite changes in pain intensity.

3.6 Conclusions

These results suggest that toe-out and trunk lean are different between those with and without knee OA. The results also suggest that patients with greater pain have greater trunk lean and contralateral pelvic drop. Despite this, these gait characteristics do not serve as acute compensatory mechanisms during a single bout of prolonged walking.

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Chapter 4:

The Effects of an Instrumented Walking Pole on Knee Joint Loading During Gait: A Technical Note

4.1 Overview

The primary objective of this study was to quantify the frontal plane forces and the moment about the knee generated by a Nordic walking pole during gait. Three dimensional gait kinematics and kinetics were collected from two healthy participants with expertise using walking poles. Walking conditions were randomized and walking speed was controlled to ±2.5% of the self-selected speed of the first condition. A compression load cell and reflective markers were applied to the pole held in the left hand. The frontal plane force and moments about the knee, including the moment created by the pole and external knee adduction moment, were primary outcome measures. Frontal plane lever arm, ground reaction forces, pole force and angles at first and second peak knee adduction moment were also assessed. At the time of first and second peak knee adduction moment, 85 and 69% of pole force acted in the frontal plane respectively. The pole provided an average moment of 1.14±0.51 and 0.10±0.09 %BW*Ht at the same points. The knee adduction moment increased 31 and 43% at first and second peak knee adduction moment respectively. The increase in first peak knee adduction moment coincided with increased frontal plane lever arm and ground reaction force. The increase in second peak knee adduction moment coincided with, increased lever arm and medial ground reaction force. Findings from these participants suggest that only a percentage of force applied to the pole is directed in the frontal plane. Despite the frontal plane force and moment provided by the pole about the knee, walking poles did not decrease knee joint loading in the individuals assessed.

4.2 Introduction

High knee joint loading has been identified as a biomechanical risk factor for the progression of medial compartment knee osteoarthritis (OA) (Miyazaki, Wada, Kawahara, Baba & Shimada, 2002). Interventions are suggested, in part, to address high knee joint loads that are often associated with knee OA. Conservative interventions that address symptoms such as activity related pain are recommended as first line treatment strategies (Hunter & Felson, 2006; Zhang et al, 2010).Nordic walking poles, also known as hiking or trekking poles, may be of interest to patients with knee OA as they are promoted by manufacturers to be a possible method to increase fitness while decreasing stress on lower extremity joints. Walking poles differ from other assistive devices such as canes. Walking poles are typically held bilaterally and are intended to promote increased walking speed and energy expenditure (Willson, Torry, Decker, Kernozek, & Steadman, 2009; Hansen & Smith, 2009) while canes are typically held unilaterally and are intended to decrease load on the knee and hip. The potential use of walking poles as a method by which patients with knee OA can reduce knee joint loading warrants further evaluation. Specifically, an analysis determining the directional components of force applied to the pole and the moment that the pole contributes about the knee is required.

Previous findings suggest that the external knee adduction moment is a reliable (Birmingham, Hunt, Jones, Jenkyn, & Giffin, 2008) and valid (Hurwitz, Sumner, Case, Block, & Andriacchi, 1998; Jackson et al., 2004; Thorp et al., 2006; Wada et al., 2001) proxy for medial compartment knee joint load. The knee adduction moment is the tendency of the tibia to rotate in the frontal plane relative to the femur due to external forces during stance. The knee adduction moment is primarily a function of the frontal plane ground reaction force and its lever arm about the knee. Applying force through a device in contact with the ground creates its own equal and opposite reaction force. When held on the contralateral side to the stance limb, the frontal plane force generates a moment that tends to abduct the femur relative to the tibia, thereby opposing the knee adduction moment (Gross and Hillstrom, 2009). Gross and Hillstrom (2009) suggest this moment is an important measure in estimating the effectiveness of walking devices in relieving the medial compartment of high loads.

Previous investigations that report effects of walking poles on the knee adduction moment vary considerably (Walter, D'Lima, Colwell, & Fregly, 2010; Stief et al., 2008; Jensen et al., 2010). Walter et al. (2010) found a decrease in knee adduction moment in one patient with an instrumented total knee replacement. However, Stief et al. (2008) found an increase in first peak knee adduction moment in 15 experienced walking pole users. Jensen et al. (2010) found no change in 10 healthy individuals applying different magnitudes of force to the poles. It is possible that the variation in previous findings measuring the knee adduction moment during pole walking may be due to the way participants use the poles. Previous studies have also reported peak axial force applied to the pole (Schiffer, Knicker, Dannohl, & Struder, 2009; Jensen et al., 2010). However, Jensen et al. (2010) suggest since poles are usually held on angle in the sagittal plane, the position and force applied to the pole could contribute to the effectiveness of the device on frontal plane moments about the knee. Therefore, quantification of directional forces and the moment provided by the pole may be useful in estimating the effectiveness of walking poles on knee joint loading.

The effect of walking poles on joint loading, particularly the contribution from the poles themselves, is currently unclear. We are unaware of any investigation that has quantified the directional components of force applied to the pole or the moment that the pole contributes about the knee. The objective of this study was to (1) quantify the frontal plane forces and moment provided by the pole and (2) compare the effect of walking poles on the knee adduction moment with unaided gait.

4.3 Methods

A convenience sample of two healthy participants with expertise using walking poles (1 male) were recruited for this study. Exclusion criteria included any musculoskeletal or neurological condition likely to affect gait, unable to speak or read English, or understand informed consent. Approval was given by the institution's Research Ethics Board for Health Sciences involving human subjects and participants gave informed consent.

Table 4.1: Subject demographics including sex, age (y), height (m), mass (kg), and body mass index (kg/m^2).

	Sex	**Age (y)**	**Height (m)**	**Weight (kg)**	**BMI (kg/m^2)**
Subject 1	M	29	1.86	82.9	24.1
Subject 2	F	24	1.62	55.9	21.3

Participants underwent three-dimensional gait analysis using an eight-camera motion capture system (Eagle cameras, EvaRT system; Motion Analysis Corp., Santa Rosa, CA, USA) synchronized with a floor-mounted force plate (Model OR6; Advanced Mechanical Technology Inc., Watertown, MA, USA). Participants wore shorts and a t shirt. A 22 marker modified Helen Hayes passive reflective set with a marker placed on the right scapula was used (Kadaba et al., 1989). Four markers were placed on the medial aspects of the knees and medial maleoli. An initial static trial was collected to determine body mass and knee and ankle joint centre locations. Additional markers were then removed. The participants walked barefoot along an 8m walkway at a self-selected pace while kinetic (1200 Hz) and kinematic (60 Hz) data were collected. During 'with poles' conditions, pole length was adjusted based on manufacturers recommendations (urbanpoling.com, 2011). Proper technique was defined as (1) maintaining trunk in a 'tall' upright position, (2) walking with contralateral pole and heel contacting the ground simultaneously, (3) contacting the tips of the poles with the ground at a point just behind the contralateral heel, (4) lifting the handle of the pole to a 'handshake position' as the ground is contacted, and (5) pressing down on the handle with a loose, comfortable grip and extended elbows.

To determine the position and angle of the pole (Urban Poling Inc., B.C., Canada) carried in the left hand, additional markers were placed at the superior and inferior aspect. To determine the load being applied to the pole, the same walking pole was instrumented with a compression load cell (Model LC201-300; Omegadyne Inc., QC, Canada). The pole was cut into two pieces directly above the inferior end. One end of the pole was fastened to each side of the load cell by screws to reattach the ends. The load cell transmitted data via a telemetry system fastened below the handle. The load cell was calibrated using the lab force plate. The pole was aligned with the vertical axis of the lab while different magnitudes of force were applied. Data from the load cell and force plate were simultaneously recorded (EvaRT 4.2; Motion Analysis Corp., Santa Rosa, U.S.A.).

Walking trials were performed until five successful right foot strikes were obtained for each condition. With poles and without poles conditions were randomized and walking speed was controlled based on the self selected speed of the condition completed first (±2.5%).

Pole force in all three directions were determined by separating axial pole force measured by the load cell into its orthogonal components using the angle of the pole and assuming the force acted along the its long axis. The frontal plane component of force applied to the pole was multiplied by the perpendicular distance to the knee and normalized to participant body weight and height (Gross & Hillstrom, 2009). The knee adduction moment was calculated from kinetic and kinematic data using commercially available (EvaRT 4.2 and Orthotrak 4.2.1; Motion Analysis Corp., Santa Rosa, CA, USA) and custom post-processing and data reduction software previously described (Hunt, Birmingham, Giffin, & Jenkyn, 2006; Jenkyn, Hunt, Jones, Giffin, & Birmingham, 2008; Hunt et al., 2008). Peak magnitudes were identified for the knee adduction moment waveform during the first and second half of stance and normalized to body weight and height.

Sagittal plane pole angle was defined as the angle of a line drawn from the inferior to the superior pole marker with respect to the lab horizontal. The lab horizontal in the forward walking direction represented zero degrees. Frontal plane pole angle was defined as the angle of a line drawn from the inferior to the superior pole with respect to the lab vertical. The lab vertical represented zero degrees. Positive values indicated an angle towards while negative indicated a pole angle away from the participant. Frontal plane lever arm, and ground reaction forces were also reported at the time of first and second peak knee adduction moment. Frontal plane lever arm was calculated as the distance between the frontal plane ground reaction force vector and the centre of rotation of the knee. Specifically, the scalar projection of the vector between the centre of pressure and knee joint centre of rotation onto the ground reaction force vector was calculated (Hunt et al. 2006). The mean of ten trials, five from each participant, were compared between conditions. Observed changes in the knee adduction moment of each participant were also compared to previously reported minimum detectable changes (Birmingham et al., 2008).

4.4 Results

Demographics are presented in Table 4.1. Average walking speed with poles was 1.62 and 1.56 m/s for subject one and two respectively. Average walking speed for without poles was 1.64 and 1.56 m/s respectively. Descriptive statistics for variables measured at first and second peak knee adduction moment are presented in Table 4.2. An ensemble curve (n=2) for the abduction moment provided by the pole and the knee adduction moment for both with and without poles are illustrated in Figure 4.1.

At the time of first peak knee adduction moment, an average force of 54.3 N were applied to the pole. Of the total force applied to the pole, 85.3% travelled in the frontal plane generating a frontal plane moment about the knee of 1.14±0.51 %BW*Ht. Despite this moment, greater first peak knee adduction moment occurred when walking with poles (2.12±1.42 vs. 1.62±0.77 %BW*Ht). Increase in first peak knee adduction moment coincided with an higher magnitudes of frontal plane lever arm, vertical ground reaction force, and medial ground reaction force. At the time of second peak knee adduction moment, an average of 9.8 N were applied to the pole. Of this force, 69.4% travelled in the frontal plane which generated a moment about the knee of 0.10±0.09 %BW*Ht. The increase in second peak knee adduction moment (2.63±1.42 vs. 1.84±0.86 %BW*Ht) coincided with higher magnitudes of frontal plane lever arm, and medial ground reaction force.

For subject one, first and second peak knee adduction moment increased 0.22 and 0.20 %BW*Ht respectively compared to walking without poles. For subject two, first and second peak knee adduction moment increased 0.76 and 1.37 %BW*H respectively compared to walking without poles. Increases for subject two were greater than reported minimal detectable changes at 85% and 95% levels of confidence, respectively (Birmingham et al., 2008).

Table 4.2: Values for all kinetic and kinematic outcome measures at first and second peak knee adduction moment for both with and without poles conditions. Means (standard deviations) were calculated from ten trials, five from each participant.

	With Poles (SD)		**Without Poles (SD)**	
	1st Peak	**2nd Peak**	**1st Peak**	**2nd Peak**
Knee adduction moment (%BW*Ht)	2.12 (0.94)	2.63 (1.42)	1.62 (0.77)	1.84 (0.86)
Lever Arm (cm)	3.97 (0.99)	2.84 (2.35)	3.34 (1.29)	2.62 (0.88)
Ground Reaction Forces (BW)				
Vertical	1.19 (0.19)	1.09 (0.11)	1.03 (0.18)	1.14 (0.13)
ML	0.08 (0.02)	0.06 (0.01)	0.05 (0.02)	0.04 (0.02)
AP	-0.24 (0.06)	0.17 (0.03)	-0.13 (0.07)	0.13 (0.07)
Pole Force (N)				
Axial Force	54.3 (25.0)	9.8 (6.2)		
Frontal Plane	46.3 (24.9)	6.8 (4.9)		
AP	27.5 (9.2)	7.1 (3.9)		
ML	4.3 (3.2)	0.9 (1.0)		
Vertical	46.1 (24.8)	6.7 (4.9)		
Pole Angle (deg)				
Sagittal	55.9 (9.4)	41.9 (4.5)		
Frontal	3.8 (2.8)	2.9 (4.5)		

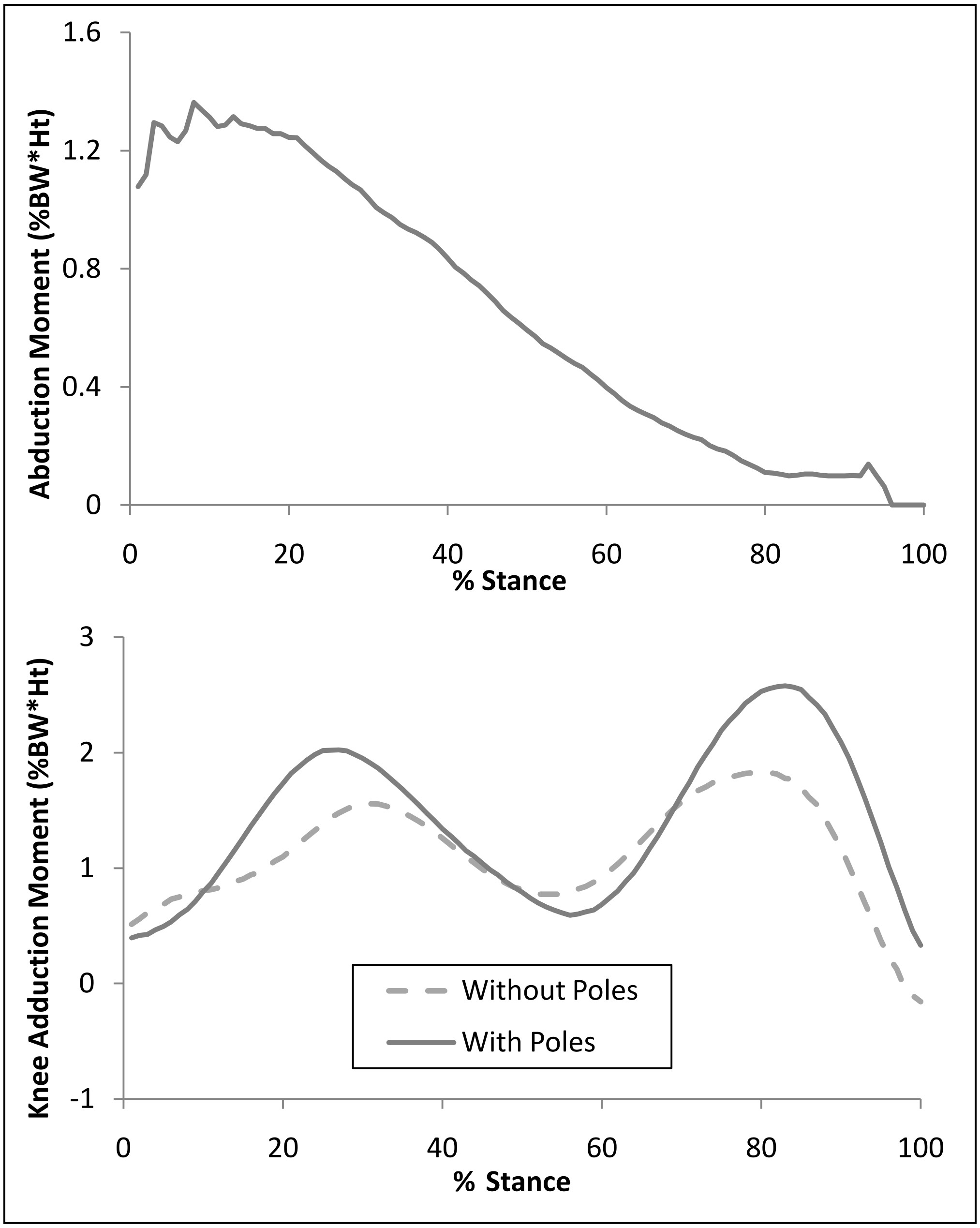

Figure 4.1. An ensemble curve (n=2) of the moment generated by the pole about the knee, and knee adduction moment, for with and without poles over 100% stance.

4.5 Discussion

Through instrumenting the pole, we were able to quantify the directional components of force as well as the moment the pole generates about the knee. We are aware of only two previous studies that measured pole force during gait. Schiffer et al. (2009) observed peak axial pole force between 36.5 and 43.3N depending on walking surface type. Jensen et al. (2010) reported peak axial force between 49 and 120N. The present findings suggest that only a portion of pole force (85.3 and 69.4%) acted in the frontal plane at first and second peak knee adduction moment. Applying a higher percentage of force in the frontal plane would have the potential of increasing the moment generated by the pole, perhaps improving the effectiveness of the device on relieving the medial compartment of the stance limb.

The use of poles lead to an increase in both first and second peak knee adduction moment respectively (31 and 43%). Increase in first peak knee adduction moment coincided with simultaneous increases in frontal plane lever arm and ground reaction force. Increase in second peak knee adduction moment coincided with an increased frontal plane lever arm and medial ground reaction force. We are aware of three previous investigations that have measured the knee adduction moment when walking with poles. Stief et al. (2008) investigated changes in knee joint loading between walking, running and walking with poles. They reported a 15% increase in first peak knee adduction moment and no significant change in second peak knee adduction moment. Walter et al. (2010) evaluated knee joint load during walking with poles. They found a decrease in both first and second peak knee adduction moment (33% and 47%) compared to walking without poles. Jensen et al. (2010) reported no change when investigating knee joint loading with increased pole force by 2.4 times. Given the present findings, variation in previous results may be in part due to the angle the pole was held and whether the peak axial force reported, coincided with the knee adduction moment peaks.

A potential limitation of this study was that the direction of force was based on the axial force applied to the pole and the direction the pole was held. It is reasonable to suggest that the true line of force did not always coincide with the long axis of the pole at first and second peak knee adduction moment. It should also be noted that pole walking tends to increase walking speed (Willson et al., 2001). By controlling for walking speed, the present increases in knee adduction moment with poles may be underestimated. Finally, although the knee adduction moment is an accepted valid and reliable measure for medial compartment knee joint load, it should be noted that the knee adduction moment does not necessarily correspond with the same result in direct medial compartment contact force (Walter et al., 2010). However, previous studies show the knee adduction moment is an appropriate non invasive, indirect measure for medial compartment knee joint load, and is highly correlated with medial compartment contact force (Zhao et al., 2007).

The present findings indicate that only a percentage of force applied to the pole is directed in the frontal plane. Despite the moment created by the force applied to the pole, walking with poles did not decrease the knee adduction moment in the individuals assessed. Altering the angle of the pole with the purpose of directing more force in the frontal plane may improve its effectiveness in altering knee joint load and should be the focus of future studies.

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Chapter 5:

The Effects of Nordic Walking Poles on the Knee Adduction Moment in Patients with Medial Compartment Knee Osteoarthritis

5.1 Overview

The objectives of this study were to evaluate the effect of walking poles on the external adduction moment about the knee during gait, and the concomitant frontal plane lever arm and vertical ground reaction force, in patients with knee OA, and (2) To explore the relationship between changes in the knee adduction moment, the force applied to the pole, and the angle at which the pole is held. Thirty-four patients with varus alignment and medial compartment knee OA underwent 3D gait analysis during trials with and without the use of poles. Conditions were randomized and walking speed during trials was controlled to ±5% of the self-selected speed of the initial condition. The pole held in the contralateral hand was instrumented with a compression load cell. Paired t-tests indicated a small, but statistically significant increase in knee adduction moment for first peak (2.88±0.79 vs. 2.71±0.78 %BW*Ht, $p=0.001$), second peak (3.05±0.85 vs. 2.88±0.82 %BW*Ht, $p<0.001$) and angular impulse (1.53±0.46 vs. 1.37±0.42 %BW*Ht*s, $p<0.001$) when using the poles. An increase in first peak knee adduction moment when using poles coincided with a significant decrease in vertical ground reaction force (0.99±0.09 vs. 1.02±0.08 BW, $p=0.015$), and an increase in lever arm (5.27±1.45 vs. 4.97±1.35 cm, $p<0.001$). Pole force in the vertical direction was inversely related ($r=-0.34$, $p=0.05$) to the increase in first peak knee adduction moment with poles. These findings suggest that despite small reductions in the vertical ground reaction force, walking with poles increase the frontal plane lever arm and therefore the knee adduction moment in patients with medial compartment knee OA. These findings also illustrate that changes are quite variable among patients, and suggest that reductions in the knee adduction moment are possible by increasing load applied to the pole in the vertical direction at the time of first peak knee adduction moment.

5.2 Introduction

The growing burden of knee osteoarthritis (OA) is now well recognized as very high by many researchers (Haq & Davatchi, 2011; Public Health Agency of Canada, 2010; Yelin, 2007). Conservative strategies are recommended as first line treatments for knee OA and include patient education, exercise and attempts to lessen aberrant knee joint loads (Hunter & Felson, 2006; Zhang et al., 2010). Nordic walking poles, also called hiking poles, may be appealing to patients with knee OA because the poles are promoted by manufactures as a method to increase fitness while reducing the stress on the joints of the lower extremities (urbanpoling.com, 2011). Walking poles are mechanically and symbolically different from other assistive devices for patients with knee OA, such as canes. Canes are typically used in one hand and are primarily intended to decrease pain by lessening the load on the hip and knee, whereas walking poles, used bilaterally in pairs, are suggested to help propel the user forward. While canes typically decrease walking speed (Chan, Smith, Kirtley, & Tang, 2005), walking poles enable increased walking speed (Willson, Torry, Decker, Kernozek, & Steadman, 2000).

There is considerable evidence to suggest that the external adduction moment about the knee is a valid (Hurwitz, Sumner, Andriacchi, & Sugar 1998; Jackson et al., 2004; Thorp et al., 2006; Wada et al., 2001), reliable (Birmingham, Hunt, Jones, Jenkyn,

& Giffin, 2008) and clinically important (Miyazaki, Wada, Kawahara, Baba, & Shimada, 2002) proxy for dynamic load on the medial compartment of the tibiofemoral joint in patients with knee OA. The knee adduction moment is calculated from 3D gait kinematics and kinetics and is determined primarily from the frontal plane lever arm and ground reaction force. Results from previous studies evaluating the effect of walking poles on the knee adduction moment vary considerably (Jensen et al., 2010; Stief et al., 2008; Walter, D'Lima, Colwell, & Fregly, 2010). Stief et al. (2008) reported an increase in the first peak knee adduction moment with the poles for 15 healthy, experienced walking pole users. Walter et al. (2010) reported a decrease in the first and second peak knee adduction moment with the poles for one subject with an instrumented total knee replacement. Jensen et al. (2010) reported no change in knee adduction moment with the poles for 10 healthy individuals applying variable loads to the poles. Importantly, patients with knee OA often have varus malalignment and exhibit higher knee adduction moments than healthy adults or those following knee joint replacement. Therefore, there may be even greater potential for walking poles to reduce the knee adduction moment in patients with knee OA with varus malalignment.

Inconsistencies in previous findings may be partly due to variability in how participants used the poles. Specifically, some individuals with knee OA may attempt to use the poles much like a cane. They may apply substantial force down through the pole (held in the hand opposite to the affected limb), thereby creating a frontal plane moment about the knee that tends to abduct the femur relative to the tibia and oppose the knee adduction moment (Gross & Hillstrom, 2009). Alternatively, other participants may apply very little force through the poles and fail to utilize their potential. Consequently, the portion of force applied through the pole in frontal plane is of particular interest. Therefore, determining the pole force in the frontal plane, the angle at which the pole is held in the sagittal plane, and their relationships with changes in the knee adduction moment may provide valuable insight.

We are unaware of any previous investigations that evaluated the effects of walking poles on the knee adduction moment in a sample of patients with knee OA, or the potential mechanisms responsible for this effect. The objectives of this study were: (1) to evaluate the effect of walking poles on the external adduction moment about the knee during gait, and the concomitant frontal plane lever arm and vertical ground reaction force, in patients with medial compartment knee OA, and (2) to explore the relationship between changes in the knee adduction moment, the force applied to the pole, and the angle at which the pole is held.

5.3 Methods

5.3.1 Participants

Thirty-four patients with knee OA (22 male) and no experience using walking poles were recruited from a tertiary care clinic specializing in orthopaedics. Patients were referred to the clinic due to knee pain and all were diagnosed with varus malalignment and OA confined mainly to the medial compartment of the tibiofemoral joint. Diagnosis of OA was based on Altman's criteria (Altman et al., 1986). Standing anteroposterior hip-to-ankle radiographs were used to measure lower limb alignment (Specogna et al., 2007) and grade severity of OA (Kellgren & Lawrence, 1957). Exclusion criteria included inflammatory or infectious arthritis of the knee, other musculoskeletal or neurological conditions likely to affect gait, unable to speak/read English, or inability to understand informed consent. Sample size was based on the ability to detect a significant within-subject difference ($p<0.05$) between conditions (with poles and without poles) of medium effect size ($d = 0.5$) 80% of the time (G*Power Version 3.1.1, Universitat Kiel, Germany). This study was approved by the institution's Research Ethics Board for Health Sciences involving human participants. All patients provided informed, signed consent.

Table 5.1: Subject demographics including sex, age (y), BMI (kg/m^2), OA grade, and mechanical axis alignment (degrees).

Age (years)	53.6 (9.8)
Height (m)	1.74 (0.10)
Mass (kg)	88.1 (16.0)
BMI* (kg/m^2)	28.3 (5.4)
Males	22
Females	12
Kellgren and Lawrence grade, number of patients	
2	17
3	12
4	4
Mechanical Axis Angle (degrees)	-6.5 (2.8)

*** BMI = Body Mass Index**

5.3.2 Gait Analysis

Patients visited the gait laboratory on two occasions. The first visit consisted of a 30 minute introductory session delivered by a trained pole walking instructor. During this session, patients were given an overview of the study, walking poles, and pole walking technique. Patients were instructed on adjusting pole length and walking technique based on the manufacturer's recommendations (urbanpoling.com, 2011). Recommendations included: (1) maintaining trunk in a 'tall' upright position, (2) walking with contralateral pole and heel contacting the ground simultaneously, (3) contacting the tips of the poles with the ground at a point just behind the contralateral heel, (4) lifting the handle of the pole to a ‘handshake position’ as the ground is contacted, and (5) pressing down on the handle with a loose, comfortable grip and extended elbows. Patients completed several practice walking trials under the guidance of the instructor until s/he felt comfortable using the poles with the described technique. After the training session was complete, patients were given access to a web-based instructional video and sent home with a set of poles for a period of one week. Patients were instructed to practice using the poles while walking and to record the frequency of use.

During the second visit to the lab, patients underwent three-dimensional gait analysis using an eight-camera motion capture system (Eagle HiRes cameras, EvaRT 4.2 system, Motion Analysis Corp., Santa Rosa, CA, USA) synchronized with a floor-mounted force plate (OR6 model, Advanced Mechanical Technology, Watertown, MA, USA). The patients wore comfortable shorts and a t-shirt. A modified Helen Hayes 22 passive reflective marker set was used that included an extra marker on the right scapula (Kadaba et al., 1989). Patients first stood motionless on the force plate to determine body mass, relative marker orientations, and the locations of the knee and ankle joint centers. Additional markers were placed on the medial aspect of the knees and medial maleoli for this static trial and were removed prior to gait testing. Patients then walked barefoot across an 8 meter walkway at a self selected speed while kinetic (1200 Hz) and kinematic (60 Hz) data were sampled.

To estimate the effectiveness of the walking poles during gait, additional markers were placed at the base of the handle and the tip of the pole. The walking pole (Urban

Poling Inc., B.C., Canada) carried in the contralateral hand to the affected limb was instrumented with a compression load cell (Model LC201-300; Omegadyne Inc., QC, Canada) and a telemetry system attached near the handle of the pole. The pole was cut into two pieces directly above the inferior end. One end of the pole was fastened to each side of the load cell by screws to reattach the ends. The load cell was calibrated using the lab force plate, by aligning the pole with the vertical axis of the lab and applying different magnitudes of force. Load cell and force plate data were synchronized and simultaneously recorded (EvaRT 4.2; Motion Analysis Corp., Santa Rosa, U.S.A.).

Patients completed walking trials until five complete force plate foot strikes of the affected limb were collected for each condition. The conditions of walking with or without poles were randomized using a random number generator conditioned for zero and one. Zero represented completing ‘with poles’ first (n=16), while one represented completing ‘without poles’ first (n=18). Walking speed was controlled to within 5% of the self-selected speed of the first condition. Pain intensity after each condition was quantified using an eleven point numerical rating scale. Patients were asked to rate the pain in their affected knee at the beginning of the visit and at the completion of each condition. Zero indicated no pain and 10 represented worst pain imaginable.

The external knee adduction moment was calculated from the kinematic and kinetic data using commercial software (Orthotrak 4.2.1; Motion Analysis Corp., Santa Rosa, U.S.A.) and custom post-processing and data reduction techniques previously described (Hunt, Birmingham, Giffin, & Jenkyn, 2006; Hunt et al., 2008; Jenkyn, Hunt, Jones, Giffin, & Birmingham, 2008). Based on the knee adduction moment waveform, we identified the peak magnitudes in the first and second halves of stance and the area under the curve (angular impulse) and normalized these values to body weight and height. We also calculated the frontal plane lever arm and vertical ground reaction force (normalized to body weight) at the time of the first and second peak knee adduction moment (Hunt et al., 2006). Frontal plane lever arm was calculated as the distance between the frontal plane ground reaction force vector and the centre of rotation of the knee. Walking speed and lateral trunk lean over the stance limb (positive angle) were also calculated because of their potential to change with the use of walking poles and their reported effect on the knee adduction moment in patients with knee OA (Hunt et al., 2008, Mundermann, Asay, Mundermann, & Andriacchi, 2008).

Pole force in the vertical direction was determined by separating axial pole force into its orthogonal components assuming force acted along the long axis of the pole. Sagittal plane pole angle was defined as the angle of a line drawn from the inferior to the superior pole marker with respect to the lab horizontal. The lab horizontal in the forward walking direction represented zero degrees. The horizontal pole force and frontal plane pole angle were also reported. Frontal plane pole angle was defined as the angle of a line drawn from the inferior to the superior pole marker with respect to the lab vertical. The lab vertical represented zero degrees. Positive values indicated an angle towards the participant while negative indicated a pole angle away from the participant. The moment created by the pole was calculated from the ground reaction force measured by the load cell. The frontal plane component of the load cell force was multiplied by the perpendicular distance between the centre of rotation of the affected knee and the pole force line of action (Gross & Hillstrom, 2009).

5.3.3 Statistical Analysis

The average of five trials for each conditions with and without poles were compared using paired t-tests and 95% confidence intervals around mean differences. The relationship between the difference in knee adduction moment, the pole force in the vertical direction, and the sagittal plane pole angle was examined using Pearson correlation coefficients. Statistical analysis was performed using the Statistical Package for the Social Sciences (SPSS v. 17; SPSS Inc., Chicago, IL).

5.4 Results

Thirty four patients volunteered and completed the study. Patient demographics are presented in Table 1. Patients reported using the poles during 43.0±22.7% of the days between initial instruction and gait analysis. Upon the second visit, walking speed with poles (1.17±0.18 m/s) was not significantly different than walking speed without poles (1.16±0.19). Patients reported an average pain intensity of 1.5±1.5 at the start of the session, 1.9±1.8 at the end of the condition with poles, and 1.8±1.8 at the end of the condition without poles.

Descriptive statistics for gait variables measured at the first and second peak knee adduction moment are presented in Tables 2 and 3, respectively. There was a significant ($p<0.05$) increase in knee adduction moment angular impulse [1.53±0.46 vs. 1.37±0.42 %BW*Ht*s, mean difference (95% CI) = 0.15 (0.09, 0.22)] with the use of poles. Ensemble average waveforms (n=34) for the knee adduction moment, frontal plane lever arm and vertical ground reaction force throughout stance during both conditions are illustrated in Figure 1. An increase in first peak knee adduction moment with poles coincided with an increase in lever arm length ($p<0.001$), a decrease in vertical ground reaction force ($p=0.015$), and a decrease in trunk lean ($p<0.001$). An increase in second peak knee adduction moment with poles coincided with no significant difference in frontal plane lever arm ($p=0.88$) and vertical ground reaction force ($p=0.19$), despite a small but statistically significant decrease in lateral trunk lean ($p=0.001$).

Change in first peak knee adduction moment was significantly correlated with the force applied through the pole in the vertical direction ($r=-0.34$, $p=0.05$) (Figure 2.), but not with pole angle in sagittal plane ($r=-0.25$, $p=0.16$). There was no significant correlation for the change in second peak knee adduction moment and the force applied through the pole in the vertical direction ($r=0.01$, $p=0.96$) or the pole angle in sagittal plane ($r=0.03$, $p=0.86$).

Table 5.2: Sample means (SD) for all kinetic and kinematic outcome measures at the time of first peak knee adduction moment for both with poles and without poles conditions and mean difference with 95% confidence intervals.

	With Poles (SD)	**Without Poles (SD)**	**Mean Difference (95% CI)**
Knee adduction moment (%BW*Ht)	2.88 (0.79)	2.71 (0.78)	0.17 (0.08,0.27)
Lever Arm (cm)	5.27 (1.45)	4.97 (1.35)	0.30 (0.15,0.44)
Trunk Lean (deg)	0.12 (1.70)	1.33 (1.65)	-1.21 (-1.59,-0.84)
Ground Reaction Forces (BW)			
Vertical	0.99 (0.09)	1.02 (0.08)	-0.02 (-0.04,-0.01)
ML	0.05 (0.01)	0.05 (0.02)	-0.003 (-0.006,0.001)
AP	-0.13 (0.05)	-0.11 (0.04)	-0.02 (-0.03,-0.01)
Pole Force (N)			
Vertical	39.3 (24.5)		
ML	4.5 (3.8)		
AP	18.0 (12.0)		
Axial Force	44.0 (26.2)		
Frontal Plane	39.6 (24.6)		
Abduction Moment (%BW*Ht)	1.25 (0.74)		
Pole Angle (deg)			
Sagittal	64.2 (11.7)		
Frontal	6.0 (3.4)		

Table 5.3: Sample means (SD) for all kinetic and kinematic outcome measures at the time of second peak knee adduction moment for both with poles and without poles conditions and mean difference with 95% confidence intervals.

	With Poles (SD)	**Without Poles (SD)**	**Mean Difference (95% CI)**
Knee adduction moment (%BW*Ht)	3.05 (0.85)	2.88 (0.82)	0.17 (0.04,0.30)
Lever Arm (cm)	4.54 (1.63)	4.56 (1.39)	-0.03 (-0.39,0.34)
Trunk Lean (deg)	0.26 (1.64)	0.99 (1.55)	-0.75 (-1.16,-0.32)
Ground Reaction Forces (BW)			
Vertical	0.99 (0.19)	1.03 (0.09)	-0.04 (-0.09,0.02)
ML	0.05 (0.02)	0.05 (0.02)	-0.003 (-0.007,0.002)
AP	0.11 (0.04)	0.10 (0.04)	0.01 (0.004,0.02)
Pole Force (N)			
Vertical	14.0 (21.0)		
ML	2.0 (3.1)		
AP	7.5 (7.5)		
Axial Force	16.1 (22.1)		
Frontal Plane	14.2 (21.1)		
Abduction Moment (%BW*Ht)	0.43 (0.54)		
Pole Angle (deg)			
Sagittal	52.9 (11.4)		
Frontal	6.7 (4.0)		

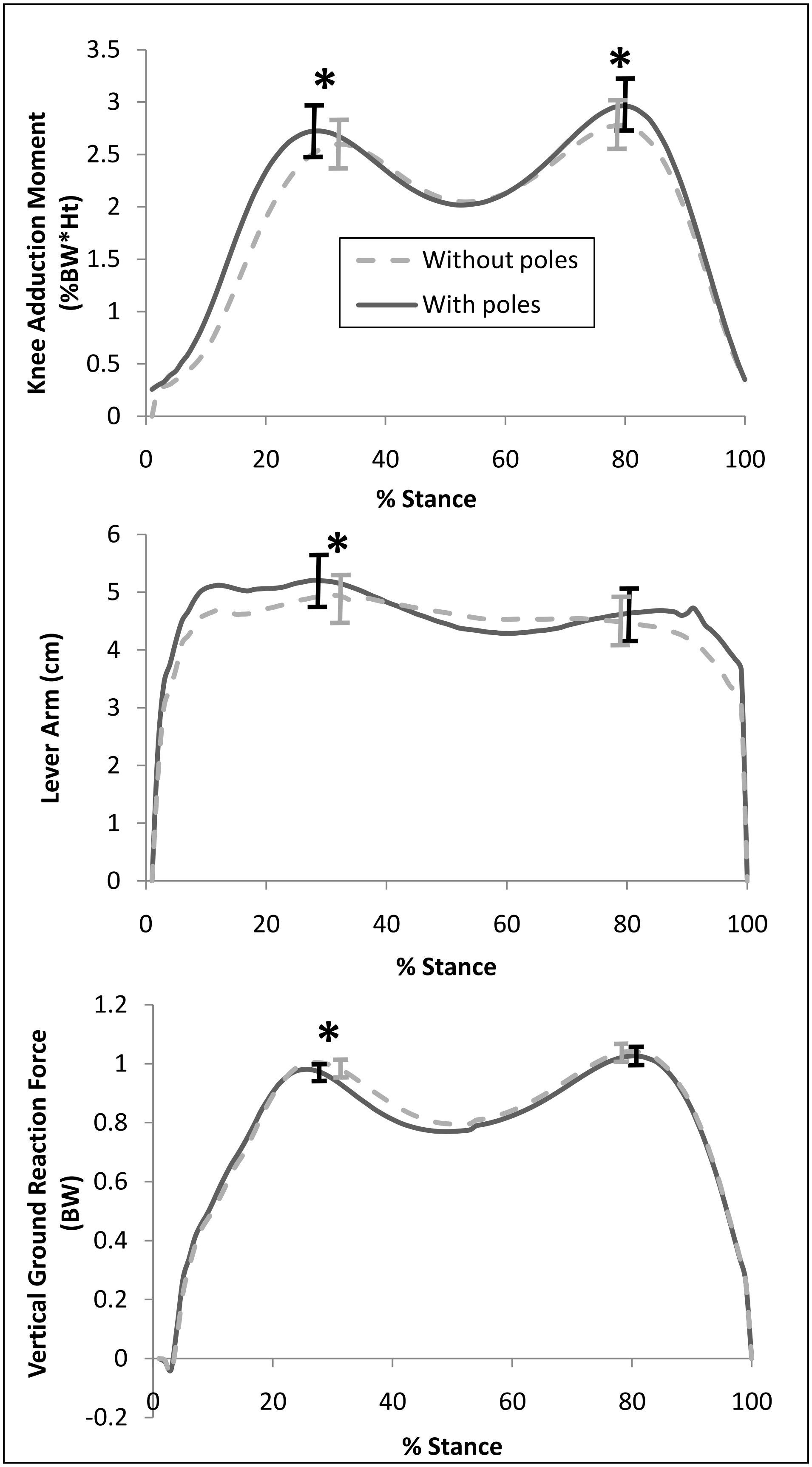
Knee Adduction Moment (%BW*Ht)
% Stance
Without poles
With poles
Lever Arm (cm)
% Stance
Vertical Ground Reaction Force (BW)
% Stance

Figure 5.1. An ensemble curve (n=34) of the knee adduction moment, frontal plane lever arm, and vertical ground reaction force over 100 percent stance with and without walking poles. 95% confidence intervals are shown for all measures at the time of first and second peak knee adduction moment. * $p<0.05$

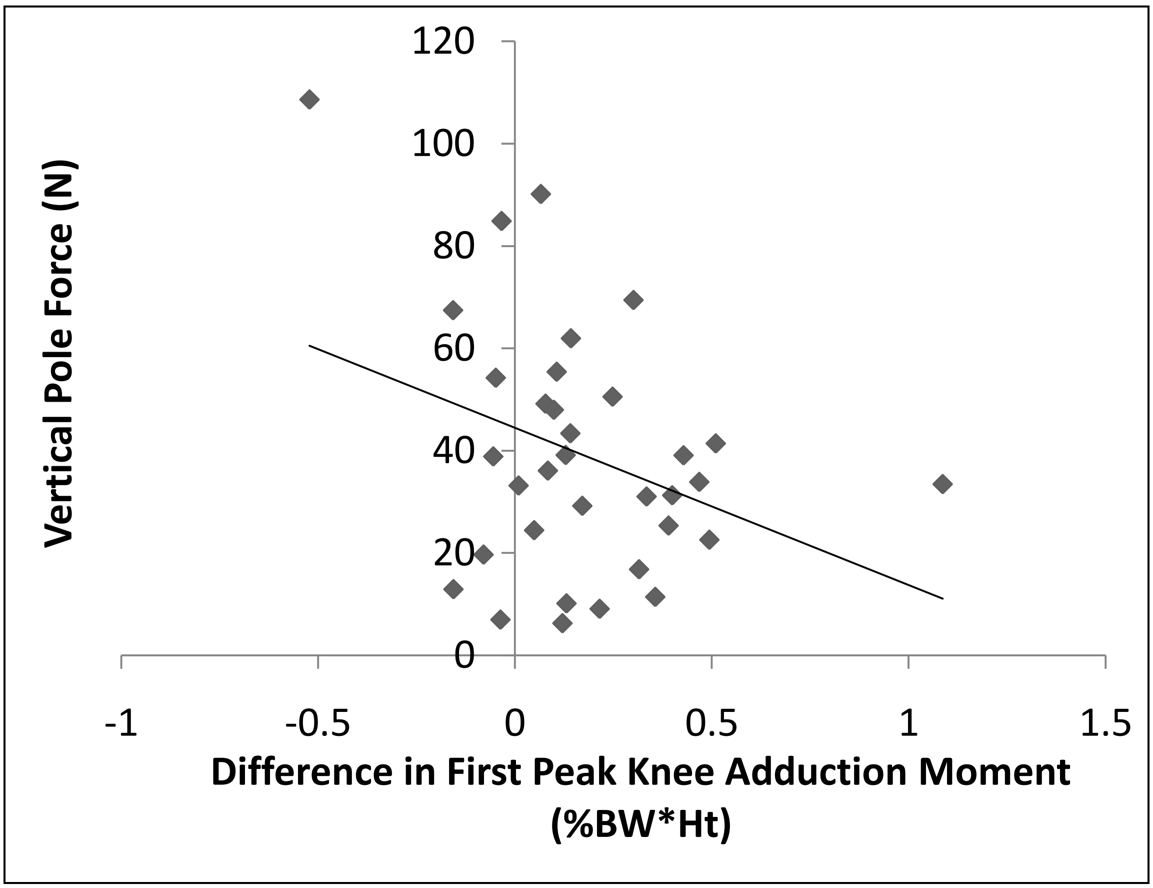

Figure 5.2. A scatter plot of the pole force in the vertical direction versus the difference in first peak knee adduction moment with and without poles. Negative values along the X axis indicated lower first peak knee adduction moment with poles. Pearson r=-0.34, p=0.05.

5.5 Discussion

Contrary to what the manufacturers propose, these findings suggest that patients with medial compartment knee OA experience an increase, rather than a decrease, in knee joint loading when using walking poles. However, it should be noted that changes were quite variable among patients, and overall, increases were quite small. Increases in the knee adduction moment first and second peak and angular impulse were 6%, 10%, and 12%, respectively. The variability in individual subject responses is consistent with the few previous investigations related to this topic. Stief et al. (2008) observed a 15% increase during first peak and no change in second peak knee adduction moment in healthy volunteers. Alternatively, Walter et al. (2010) observed a 33% and 47% decrease in the first and second peak knee adduction moment in one subject with an instrumented total knee replacement. Importantly, some subjects in the present study also experienced decreases in the knee adduction moment when using the poles (Figure 5.2), although these were not as large as in the subject evaluated by Walter et al.

Several investigators have reported the importance of frontal plane lever arm, ground reaction force and lateral trunk lean to the magnitude of knee adduction moment (Hunt et al., 2006; Hunt et al., 2008; Mundermann et al., 2008). For example, Hunt et al. (2006) investigated associations among the knee adduction moment, lever arm and frontal plane ground reaction force. They determined that frontal plane lever arm was significantly correlated with the knee adduction moment in 100 patients with medial compartment knee OA. Hunt et al. (2008) subsequently investigated the role of select gait kinematics on the knee adduction moment. They determined that 13% of the variation in first peak knee adduction moment was explained by lateral trunk lean in 120 patients with medial compartment knee OA. Additionally, Mundermann et al. (2008) investigated implications of increasing lateral trunk lean during gait on knee joint load. They determined that 10 degrees increase in trunk lean decreased the knee adduction moment by an average of 65% for 19 healthy participants. Lateral trunk lean is argued to be a compensatory mechanism for patients with knee OA to reduce high knee joint loads. Our findings suggest that walking with poles reduces this protective mechanism by encouraging less lateral trunk lean towards the stance limb. It is therefore possible that a decrease in lateral trunk lean towards the stance limb is responsible for the increase in frontal plane lever arm and consequently the knee adduction moment.

The current study determined that change in knee joint load was significantly correlated with the amount of force applied through the pole in the vertical direction, but not with sagittal pole angle during first peak knee adduction moment. Findings suggest that patients who were able to apply a greater load in the vertical direction during first peak may improve the effectiveness of the poles in altering knee joint load. For example, the greatest reduction in first peak knee adduction moment in a single patient was achieved with an axial force of 109.9N, where 98.8% of the force was traveling in the vertical direction. Conversely, the greatest increase in first peak in a single patient occurred with an axial force of 37.2N with 89.8% of force traveling in the vertical direction. Therefore, users of walking poles should attempt to apply more force directed in the vertical direction as close to first peak knee adduction moment as possible. At the time of second peak knee adduction moment an average of 87% of axial pole force was directed along the vertical axis however, no significant correlation with change in knee adduction moment was found. A lack of correlation during second peak knee adduction moment with pole force in the vertical direction is likely due to the relatively small amount of force applied at that particular time of stance.

We are aware of only two previous studies that have reported axial pole force. Schiffer, Knicker, Dannohl, and Struder (2009) investigated energy cost during pole walking. They reported peak axial pole force between 36.5 and 43.3N depending on the type of walking surface. Jensen et al. (2010) investigated the effects of different magnitudes of pole force on knee joint load. They found no change in first or second peak knee adduction moment even when pole force was increased by 2.4 times. Our findings suggest that patients applied comparable pole force of 39.3 N, however further comparison is difficult since pole kinematics were not reported by Schiffer et al. and Jensen et al. studies. Our findings are different from Jensen et al. who observed no change in knee adduction moment with increased pole force. However, the absence of change may be due to the pole being held on a steeper angle, or perhaps a difference in timing between the peak axial pole force and the first peak knee adduction moment.

A potential limitation of this study was that patients reported only mild to moderate pain. It is reasonable to suggest that individuals with severe OA symptoms may apply more force on the poles in an attempt to relieve symptoms. Additionally, patients reported using the poles only 43±22.7% of the days the poles were in the patient's possession. Additional time spent using the poles may have increased their ability to apply force and improved their efficiency. It should also be noted that pole walking tends to increase walking speed (Willson et al., 2000). We controlled walking speed to increase internal validity. By doing so the present increases in knee adduction moment with pole use may be underestimated. It should be noted that a change in knee adduction moment does not necessarily correspond with the same result in direct medial compartment contact force (Walter et al., 2010). However, previous findings show that the knee adduction moment is an appropriate non invasive technique for measuring medial compartment knee joint loading and is highly correlated with medial compartment contact force (Zhao et al., 2007).

The results from this study suggest that walking pole users must weigh the cost of marginally increasing knee joint loading, with the benefits of previously reported increased mobility, and possible improvements in independence, function, and safety (Oakley, Zwierska, Tew, Beard, & Saxton, 2008; Willson et al., 2000; Van der Esch, Heijmans, & Dekker, 2003). Future research should evaluate variations in pole walking technique, such as timing of pole loading with respect to knee loading, in order to determine an optimal technique for unloading the medial compartment of the knee.

5.6 Conclusions

Findings of this study suggest that despite small reductions in the vertical ground reaction force, walking with poles increases the frontal plane lever arm and therefore the knee adduction moment in patients with medial compartment knee OA. These findings also illustrate that changes are quite variable among patients, and suggest that reductions in the knee adduction moment are possible by increasing load applied to the pole in the vertical direction at the time of first peak knee adduction moment.

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Chapter 6:

Discussion

6.1 Overview

The overall objective of this thesis was to explore the effects of prolonged walking and the use of Nordic walking poles on selected gait characteristics indicative of knee joint load. Main findings included:

Chapter 2: Maximum lateral trunk lean angle and maximum toe-out angle demonstrated consistent behaviour (i.e. did not change over time) during prolonged walking in healthy adults. These measures also demonstrated moderate to good test-retest reliability (ICCs ranged from 0.61 to 0.95) and concurrent validity (ICCs ranged from 0.88 to 0.92). This study confirmed that lateral trunk lean and toe-out angles during prolonged bouts of treadmill walking are adequate for studying potential compensatory gait mechanisms.

Chapter 3: Maximum toe-out angle was smaller in patients with knee OA than healthy adults [mean difference (95% CI) = 3.2 (-0.4, 6.7) to 4.0 (0.5, 7.4) degrees] and changed over time, but not in a systematic way. Maximum lateral trunk lean angle was higher in patients with knee OA than healthy adults [mean difference (95% CI) = -1.4 (-2.4, -0.3) to -1.1 (-2.2, 0.1) degrees] but did not change over time. Pelvic rise was also higher in patients with knee OA than healthy adults [mean difference (95% CI) = -1.9 (-3.1, -0.6) to -1.2 (-0.3, 0.2) degrees], but did not change over time. Importantly, trunk lean and toe-out did not increase during prolonged walking in accordance with increases in pain intensity. Patients with greater pain intensity did have greater trunk lean (r = 0.65 to 0.64) and contralateral pelvic drop (r = 0.47) than other patients with more mild pain intensities. These findings suggest that despite a significant increase in pain intensity, maximum toe-out and trunk lean angle did not serve as acute compensatory gait mechanisms.

Chapter 4: Through instrumentation of a walking pole, findings indicated that only a portion of force applied to the pole acted in the frontal plane. Despite the frontal plane moment that was generated by the force applied to the pole, the knee adduction moment increased at both first and second peak knee adduction moment. Applying force to the pole while being held on an angle possibly limited the effectiveness of the pole in reducing knee joint load.

Chapter 5: The use of walking poles in patients with medial compartment knee OA did not result in reductions in the knee adduction moment. In fact, despite a significant decrease in vertical ground reaction force [mean difference (95% CI) = -0.02 (-0.04,-0.01)], the use of poles coincided with a significant increase in lever arm length [mean difference (95% CI) = 0.30 (0.15,0.44)], possibly due to a significant decrease in lateral trunk lean towards the stance limb [mean difference (95% CI) = -1.21 (-1.59,-0.84)], and resulted in a significant increase in the knee adduction moment [mean difference (95% CI) = 0.17 (0.08,0.27)]. Pole force in the vertical direction was inversely related (r = -0.34) to the increase in first peak knee adduction moment with poles. Changes were quite variable among patients and suggested that reductions in the knee adduction moment are possible by increasing load applied to the pole in the vertical direction at the time of first peak knee adduction moment.

6.2 Implications and Future Research

Compensatory gait mechanisms (Chang 2007; Mundermann 2005; Wang, Kuo, Andriacchi, & Galante, 1990) and several therapeutic devices (Lindenfeld, Hewett, & Andriacchi, 1997; Kerrigan, Lelas, Goggins, Merriman, Kaplan, & Felson, 2002; Fregly, D'Lima, & Colwell, 2009; Oakley, Zwierska, Tew, Beard, & Saxton, 2008) have been reported to reduce dynamic knee joint load in both healthy adults and patients with knee OA. Consistent with previous investigations, current findings demonstrated that those with knee OA tend to have greater trunk lean and less toe-out than their healthy counterparts (Lynn & Costigan, 2008; Hunt, Wrigley, Hinman, & Bennell, 2010). This is of importance since greater toe-out angles have been associated with lower risk of OA progression (Chang et al., 2007). Patients in this thesis experienced an increase in pain intensity over the prolonged period of treadmill walking; however, trunk lean and toe-out measures remained consistent. Those with high pain intensity had greater trunk lean angles; however, neither trunk lean or toe-out acted as acute compensatory mechanisms during a single bout of prolonged walking. Future studies should investigate trunk lean and toe-out using prospective study designs over longer periods of time in patients with knee OA to better understand potential gait compensations.

In a related manner, standard errors of measurement were also assessed for trunk lean and toe-out in this thesis. These findings suggest that although these measures may be adequate to determine differences between groups, care must be taken when prospectively evaluating individual patient changes.

Patients with knee OA often seek other forms of intervention to address their unresolved symptoms. As a result, walking poles may be an attractive option for patients as they are suggested to help reduce the stress on the load bearing joints of the lower extremity. Findings from previous studies vary considerably regarding their effects on knee joint loading (Fregly et al. 2009; Walter, D'Lima, Colwell, & Fregly, 2010; Jensen, Henriksen, Aaboe, Hansen, Simonsen, & Alkjaer, 2010). Findings from this thesis suggest instrumentation of a walking pole provided valuable information regarding the angle the pole was held and the direction of force applied to the pole during gait. Furthermore, walking with poles using the manufacturer recommended technique marginally increased (rather than decreased) the knee adduction moment in patients with knee OA. It is important to note that an increase in knee adduction moment was not observed for all individuals. For those where the knee adduction moment did not increase, and in some instances decreased, greater force was applied through the pole in the vertical direction.

As the poles were often held on an angle in the sagittal plane, a higher percentage of force in the vertical direction may improve the poles effectiveness in reducing knee joint loading. Further research should determine the role of variations in pole walking technique, such as timing and direction of peak pole force with respect to the knee adduction moment in patients with knee OA.

6.3 Walk Softly and Carry a Big Stick: Training Required

The present findings suggest that although biomechanically plausible, patients with knee OA do not appear to naturally adopt gait characteristics, or use walking poles in response to symptoms such as pain, or in a way to decrease knee joint loads. It may therefore be important that individuals receive training to properly adapt technique to maximize the potential benefits of compensatory mechanisms and therapeutic devices. These findings support previous calls for the analysis of gait retraining as possible therapeutic interventions for patients with knee OA (Chang et al., 2007; Mundermann, Asay, Mundermann, & Andriacchi, 2008; Guo, Axe, & Manal, 2007). Recent feasibility studies have suggested reductions in knee joint loading are possible in healthy adults (Barrios, Crossley, & Davis, 2010; Hunt, Simic, Hinman, Bennell, & Wrigley, 2011). However, difficulty and discomfort in performing the requested gait patterns have also been reported.

In the present thesis, immediate changes in the primary outcome measure that would be consistent with decreases in knee joint load were hypothesized. Specifically, in chapter three, a progressive increase in trunk lean and toe-out was anticipated in response to increased levels of pain. In chapter four and five, a decrease in knee adduction moment was anticipated with the use of poles. For both prolonged walking and walking with poles, these hypotheses were not met. Thus, immediate instruction and/or biofeedback could be of great importance in these areas.

6.4 Limitations

Important limitations to this series of studies should be acknowledged. For instance, this thesis investigated the test-retest reliability and concurrent validity of trunk lean and toe-out in healthy adults. Past studies have investigated test-retest reliability of trunk lean and toe-out for patients with knee OA during over ground gait analysis (Hunt et al., 2008). A post hoc analysis using data collected in chapter three indicated good concurrent validity for trunk lean and toe-out during treadmill walking for individuals with knee OA. However, more work is required to fully understand day to day reliability of treadmill walking for patients with knee OA. It is also regrettable that we were unable to measure the knee adduction moment during prolonged treadmill walking. Limitations associated with the equipment only allowed for the collection of select kinetics.

Limitations also existed for studies investigating walking poles. Patients reported using the poles less than half of the days the poles were in the patient's possession. Additional time spent using the poles may have increased patients' ability to apply force and may have improved their efficiency. However, it should also be noted that walking with poles tends to increase walking speed (Willson, Torry, Decker, Kernozek, & Steadman, 2000) and that the knee adduction moment increases with speed. We controlled walking speed (i.e. kept the speed the same during conditions with and without poles) to increase the internal validity of our comparisons. By doing so, the observed increases in knee adduction moment with pole use may actually underestimate true increases that may occur with increases in speed.

On average, patients with knee OA examined in this thesis experienced only mild pain. It is plausible that patients experiencing more pain may rely on compensatory gait mechanisms more, or use walking poles differently, to lessen the load on their symptomatic knee. In addition, numbers of men and women were not equal in chapters 2, 3, or 5. Having more women in chapters 2 and 5, or matched controls in chapter 3, would make the studies more generalizable to the greater population of persons with knee OA.

Finally, although the knee adduction moment is a valid (Hurwitz, Sumner, Case, & Block, 1998; Jackson et al., 2004; Thorp et al., 2006; Wada et al., 2001) and reliable (Birmingham, Hunt, Jones, Jenkyn, & Giffin, 2008) proxy for joint load, with accepted clinical significance (Miyazaki, Wada, Kawahara, Baba, & Shimada, 2002), a change in knee adduction moment does not necessarily correspond with the same result in direct medial compartment contact force (Walter et al., 2010). However, previous findings show that the knee adduction moment is an appropriate non-invasive technique for measuring medial compartment knee joint loading that is highly correlated with medial compartment contact force (Zhao et al., 2007).

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Appendix A: Letter of Information and Consent Forms

LETTER OF INFORMATION

Investigators: Daniel Bechard, Trevor Birmingham, Robert Giffin, Ian Jones, Tom Jenkyn

Project Title: **A pilot study of gait compensations in individuals with knee osteoarthritis.**

What is the purpose and what are the potential benefits of the study?

The purpose of this study is to observe variations in walking over a prolonged period of time in individuals with knee osteoarthritis.

What are the criteria for participating in the study?

You are invited to participate in this study because you have osteoarthritis of the knee, are between the ages of 30 and 70 and have no other known conditions that might affect walking.

What is the procedure?

If you agree to participate, you will be asked to perform several walking trials in the Wolf Orthopaedic Biomechanics Lab in the Fowler Kennedy Sport Medicine Clinic at the 3M Centre, University of Western Ontario. You will be asked to complete six to ten walking trials over a ten meter runway. You will then be asked to walk on a treadmill at the same pace for 30 minutes or until you, the participant, terminate the protocol. During this time, your pain levels will be continually monitored. A safety harness will be offered to you if you so desire. This harness is industrial grade and linked to the ceiling via safety straps. We encourage you to approach all walking trials as you would in a normal, everyday setting. Standard of care for individuals with knee osteoarthritis includes gait analysis and your participation in this study will have no effect on your standard of care. Collecting information regarding prolonged walking will be completed in addition to this standard. In order to complete this collection, reflective markers will be placed over your toe, heel, ankle, knee, pelvis, scapula, shoulders, elbows, and wrists allowing the analysis of your movements during walking. These will be placed on you using two sided hypoallergenic tape. High resolution cameras will record the motion of these markers as you walk through the laboratory. Although we encourage you to bring your own shorts, we can provide this for you.

How long and how many visits does the testing involve?

The testing will be completed within one test session. We anticipate 60 minutes to allow for warm-up, and completion of the tests.

Are there any discomforts or risks associated with testing?

There are no identified risks in participating in this study beyond the normal risks associated with walking over ground and on a treadmill. We will continually monitor any existing pain throughout the session and you may terminate the protocol at any point. If you take pain or arthritic medication you are still eligible to participate in this study. You will not be asked to refrain, delay, or adjust your medication dosages at any time. As a participant to the study parking will be covered.

Will the results be kept confidential?

Your individual results will be held in strict confidence. No person other than the investigators will be given access to your records without your expressed permission. When the results are reported, individual records will be coded or reported as group data. Representatives of The University of Western Ontario Health Sciences Research Ethics Board may contact you or require access to your study-related records to monitor the conduct of the research.

It is the right of the participant to decide independently whether or not they would like to be able to review the results independently. No identifiable information will be shared with any other individuals. If you would like a copy of the study results, please indicate this and your contact information on a separate page.

Is your participation voluntary?

Participation in the study is voluntary. You may refuse to participate or withdraw from the study at any time with no effect on your current or future care. Refusal to participate in the study will have no effect on your delivery of healthcare. Participation in this study does not prevent you from participating in other research studies at the present time or in the future. If you choose to withdraw at any point, it is your option whether previously collected data can be used by the researchers or destroyed. No other researchers will have access to the data collected in this project beyond the ones listed below.

How long will data be stored? Will I be called upon in the future?

The investigative team would like for data to be retained indefinitely to facilitate future investigations within this area of study. This is voluntary and you may refuse the retention of this information, requesting that your data and identifying information such as your name may be disposed of once the current line of research is completed. If you are willing to be contacted in the future for other research studies, please indicate this on the consent form. You can always withdraw this consent to be contacted in the future, should you change your mind. In this case, your name and contact information will be removed from our records. You may decline being contacted for further research that may continue from this project.

Who should you contact with any questions?

Please contact us at the address below, or by phone, to ask any questions you may have about the study.

Daniel Bechard BA, MSc
Graduate Student
Faculty of Health Sciences-Rehabilitation Science
The University of Western Ontario
Wolf Orthopaedic Biomechanics Lab
Fowler Kennedy Sport Medicine Clinic
London, Ontario, N6A 3K7

Trevor Birmingham PhD
Associate Professor
Wolf Orthopaedic Biomechanics Lab
Fowler Kennedy Sport Medicine Clinic
London, Ontario, N6A 3K7

If you have any questions about your rights as a research participant or the conduct of the study you may contact, Dr. David Hill, Scientific Director, Lawson Health Research Institute.

Please keep this information letter for future reference.

CONSENT FORM

Gait Compensations in Individuals with Knee Osteoarthritis

I have read the Letter of Information, have had the nature of the study explained to me and I agree to participate. All questions have been answered to my satisfaction.

_____________________	_______________________	________________
Print Name	Signature	Date

Signature of Person Obtaining Consent

_______________________	_______________________	________________
Print Name	Signature	Date

6.5.1 Possibility of future research

There may be future opportunities for you to participate in ongoing research. If you are interested in being contacted, please check the appropriate box below. If contacted, you will be asked to read a new letter of information and sign a new consent form.

□ Please do not keep my name and contact information. I do not wish to be contacted in the future.

□ Please keep my name and contact information so that I may be contacted to learn about future research opportunities or have access to my data in the future.

Copy of Study Results

I would like a copy of the study results. Yes □ No □

If yes, please write your mailing address below.

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LETTER OF INFORMATION
Investigators: Daniel Bechard, Robert Giffin, Tom Jenkyn, Trevor Birmingham

Project Title: **The effects of Nordic walking poles on knee joint loading for people with knee osteoarthritis**

What is the purpose and what are the potential benefits of the study?

This study will investigate whether or not Nordic walking poles decrease the weight bearing force on the knee during walking. This study will add to our understanding of potential treatments for patients with knee osteoarthritis.

What are the criteria for participating in the study?

You are invited to participate in this study because you have osteoarthritis of the knee, are between the ages of 30 and 70 and have no other known conditions that might affect walking. No previous experience with Nordic walking is necessary.

What is the procedure?

If you agree to participate, you will be introduced to Nordic walking by a trained instructor. You will be given a set of Nordic walking poles to take home and practice with. You will be asked to use these poles in environments where you feel comfortable to do so. Walking outside, such as in parks, on sidewalks or in open spaces, are ideal areas. However, if you feel comfortable with using these devices during everyday tasks completed indoors, you are welcome to do so. The researcher's only requirement is that you feel comfortable using the poles upon the day of data collection in our lab. Upon your return approximately 5 days later but upon your schedule, you will be asked to perform several walking trials in the Wolf Orthopaedic Biomechanics Lab in the Fowler Kennedy Sport Medicine Clinic at the 3M Centre, University of Western Ontario. You will be asked to complete twenty to thirty walking trials over a ten meter runway with and without Nordic walking poles. We encourage you to approach all walking trials as you would in a normal, everyday setting. Reflective markers will be placed over your toe, heel, ankle, knee, pelvis, scapula, shoulders, elbows, and wrists allowing the analysis of your movements during walking. Standard of care for individuals with knee osteoarthritis includes gait analysis and your participation in this study will have no effect on your standard of care. Collecting information regarding individuals walking with poles will be completed in addition to this standard. In order to complete this collection, reflective markers will be placed over your toe, heel, ankle, knee, pelvis, scapula, shoulders, elbows, and wrists allowing the analysis of your movements during walking.

These will be placed on you using two sided hypoallergenic tape. High resolution cameras will record the motion of these markers as you walk through the laboratory. Although we encourage you to bring your own shorts, we can provide this for you.

How long and how many visits does the testing involve?

The testing will be completed within one test session. We anticipate the initial introductory visit will take approximately 30 minutes. The data collection visit (second visit) will take 30 minutes to one hour of time to allow for warm-up, and completion of the tests.

Are there any discomforts or risks associated with testing?

There are no identified risks in participating in this study beyond the normal risk of pain you may experience during walking. As a participant to the study parking will be covered.

Will the results be kept confidential?
Your individual results will be held in strict confidence. No person other than the investigators will be given access to your records without your expressed permission. When the results are reported, individual records will be coded or reported as group data. Representatives of The University of Western Ontario Health Sciences Research Ethics Board may contact you or require access to your study-related records to monitor the conduct of the research.

It is the right of the participant to decide independently whether or not they would like to be able to review the results independently. No identifiable information will be shared with any other individuals. If you would like a copy of the study results, please indicate this and your contact information on a separate page.

Is your participation voluntary?

Participation in the study is voluntary. You may refuse to participate or withdraw from the study at any time with no effect on your current or future care. Refusal to participate in the study will have no effect on your delivery of healthcare. Participation in this study does not prevent you from participating in other research studies at the present time or in the future. If you choose to withdraw at any point, it is your option whether previously collected data can be used by the researchers or destroyed. No other researchers will have access to the data collected in this project beyond the ones listed below.

How long will data be stored? Will I be called upon in the future?

The investigative team would like for data to be retained indefinitely to facilitate future investigations within this area of study. This is voluntary and you may refuse the retention of this information, requesting that your data and identifying information such as your name may be disposed of once the current line of research is completed. If you are willing to be contacted in the future for other research studies, please indicate this on the consent form. You can always withdraw this consent to be contacted in the future, should you change your mind. In this case, your name and contact information will be removed from our records. You may decline being contacted for further research that may continue from this project.

Who should you contact with any questions?

Please contact us at the address below, or by phone, to ask any questions you may have about the study.

Daniel Bechard BA, MSc
Graduate Student
Faculty of Health Sciences-Rehabilitation Science
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Trevor Birmingham PhD
Associate Professor
Wolf Orthopaedic Biomechanics Lab
Fowler Kennedy Sport Medicine Clinic
London, Ontario, N6A 3K7

If you have any questions about your rights as a research participant or the conduct of the study you may contact, Dr. David Hill, Scientific Director, Lawson Health Research Institute.

Please keep this information letter for future reference.

CONSENT FORM
The effects of Nordic walking poles on knee joint loading for people with knee OA

I have read the Letter of Information, have had the nature of the study explained to me and I agree to participate. All questions have been answered to my satisfaction.

____________________	____________________	________________
Print Name	Signature	Date

Signature of Person Obtaining Consent

____________________	____________________	________________
Print Name	Signature	Date

*6.5.2 **Possibility of future research***

There may be future opportunities for you to participate in ongoing research. If you are interested in being contacted, please check the appropriate box below. If contacted, you will be asked to read a new letter of information and sign a new consent form.

□ Please do not keep my name and contact information. I do not wish to be contacted in the future.

□ Please keep my name and contact information so that I may be contacted to learn about future research opportunities or have access to my data in the future.

Copy of Study Results

I would like a copy of the study results. Yes □ No □

If yes, please write your mailing address below.

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Appendix B: Ethics Approval

Office of Research Ethics
The University of Western Ontario
Room 4180 Support Services Building, London, ON, Canada N6A 5C1
Telephone: (519) 661-3036 Fax: (519) 850-2466 Email: ethics@uwo.ca
Website: www.uwo.ca/research/ethics

Use of Human Subjects - Ethics Approval Notice

Principal Investigator: Dr. T.B. Birmingham
Review Number: 16075E **Review Level:** Expedited
Review Date: April 08, 2009
Protocol Title: A pilot study of gait compensations in individuals with knee osteoarthritis
Department and Institution: Physical Therapy, University of Western Ontario
Sponsor: INTERNAL RESEARCH FUND-UWO
Ethics Approval Date: April 28, 2009 **Expiry Date:** May 31, 2010
Documents Reviewed and Approved: UWO Protocol, Letter of Information and Consent
Documents Received for Information:

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced study on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the UWO Updated Approval Request Form.

During the course of the research, no deviations from, or changes to, the protocol or consent form may be initiated without prior written approval from the HSREB except when necessary to eliminate immediate hazards to the subject or when the change(s) involve only logistical or administrative aspects of the study (e.g. change of monitor, telephone number). Expedited review of minor change(s) in ongoing studies will be considered. Subjects must receive a copy of the signed information/consent documentation.

Investigators must promptly also report to the HSREB:

a) changes increasing the risk to the participant(s) and/or affecting significantly the conduct of the study;
b) all adverse and unexpected experiences or events that are both serious and unexpected;
c) new information that may adversely affect the safety of the subjects or the conduct of the study.

If these changes/adverse events require a change to the information/consent documentation, and/or recruitment advertisement, the newly revised information/consent documentation, and/or advertisement, must be submitted to this office for approval.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

[redacted]

Chair of HSREB: Dr. Joseph Gilbert

[redacted]

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LHRI

UWO HSREB Ethics Approval - Initial
V.2008-07-01 (rptApprovalNoticeHSREB_Initial) 16075E Page 1 of 1

Office of Research Ethics

The University of Western Ontario
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Telephone: (519) 661-3036 Fax: (519) 850-2466 Email: ethics@uwo.ca
Website: www.uwo.ca/research/ethics

Western

Use of Human Subjects - Ethics Approval Notice

Principal Investigator: Dr. T.B. Birmingham
Review Number: 15792E **Review Level:** Expedited
Review Date: January 14, 2009
Protocol Title: The effects of Nordic walking poles on knee joint loading for people with knee osteoarthritis
Department and Institution: Physical Therapy, University of Western Ontario
Sponsor:
Ethics Approval Date: March 18, 2009 **Expiry Date:** December 31, 2009
Documents Reviewed and Approved: UWO Protocol, Letter of Information and Consent. Advertisement.
Documents Received for Information:

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced study on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the UWO Updated Approval Request Form.

During the course of the research, no deviations from, or changes to, the protocol or consent form may be initiated without prior written approval from the HSREB except when necessary to eliminate immediate hazards to the subject or when the change(s) involve only logistical or administrative aspects of the study (e.g. change of monitor, telephone number). Expedited review of minor change(s) in ongoing studies will be considered. Subjects must receive a copy of the signed information/consent documentation.

Investigators must promptly also report to the HSREB:

a) changes increasing the risk to the participant(s) and/or affecting significantly the conduct of the study;
b) all adverse and unexpected experiences or events that are both serious and unexpected;
c) new information that may adversely affect the safety of the subjects or the conduct of the study.

If these changes/adverse events require a change to the information/consent documentation, and/or recruitment advertisement, the newly revised information/consent documentation, and/or advertisement, must be submitted to this office for approval.

Members of the HSREB who are named as investigators in research studies, or declare a conflict of interest, do not participate in discussion related to, nor vote on, such studies when they are presented to the HSREB.

[redacted]

Chair of HSREB: Dr. Joseph Gilbert

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Use of Human Subjects - Ethics Approval Notice

Principal Investigator: Dr. T.B. Birmingham
Review Level: Expedited
Review Number: 15792E
Revision Number: 1
Review Date: April 7, 2010
Approved Local # of Participants: 34
Protocol Title: The effects of Nordic walking poles on knee joint loading for people with knee osteoarthritis
Department and Institution: Physical Therapy, University of Western Ontario
Sponsor:
Ethics Approval Date: April 7, 2010
Expiry Date: April 30, 2011
Documents Reviewed and Approved: Revised Study End Date
Documents Received for Information:

This is to notify you that The University of Western Ontario Research Ethics Board for Health Sciences Research Involving Human Subjects (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices: Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced revision(s) or amendment(s) on the approval date noted above. The membership of this REB also complies with the membership requirements for REB's as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB's periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the UWO Updated Approval Request Form.

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Chair of HSREB: Dr. Joseph Gilbert
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	Province of Ontario Graduate Scholarship in Science and Technology 2009-2010
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	Rowing Coach Rowing Canada, the University of Western Ontario 2006-2011

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Bechard, D., Birmingham, T., Zecevic, A., Jenkyn, T. (2011). Time-varying behavior, test-retest reliability and concurrent validity of lateral trunk lean and toe-out angles during prolonged treadmill walking. *Gait and Posture*, *34*(1), 81-85.

Bechard, D., Nolte, V., Kedgley, A., & Jenkyn, T. (2009). Total kinetic energy production of body segments is different between racing and training pace in elite Olympic rowers. *Sports Biomechanics*, *8*(3), 199-211.