KNEE BIOMECHANICS WALKING ON RAILROAD BALLAST AND
THE ASSOCIATED RISK FACTORS FOR KNEE OSTEOARTHRITIS

by

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This thesis has been read by each member of the following supervisory committee and by majority vote has been found to be satisfactory.

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2010
To the Graduate Council of the University of Utah:

I have read the thesis and have found that (1) the content is consistent and acceptable; (2) its illustrative materials including figures, tables, and charts are in place; and (3) the final manuscript is satisfactory to the supervisory committee and is ready for submission to The Graduate School.

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ABSTRACT

Walking is an important part of everyday life. Osteoarthritis (OA) is a musculoskeletal disorder affecting approximately 20% of adults. The knee is the joint most commonly affected by OA. Knee OA can create severe functional limitations and a decreased quality of life. Walking on crushed rock aggregate is a common and necessary part of the duties of some railroad workers. There is little research describing the increased risks for OA that could be associated with walking on crushed rock aggregate. To measure the external biomechanics at the knee two adjustable tracks were constructed, one with main-line (large) ballast and the other with yard (small) ballast. One of the tracks was modified to represent a normal walking surface. The tracks had an adjustable transverse plane slope. Two slope conditions were used, normal flat condition and 7° sloped condition. Data were collected from 10 railroad workers walking on three surfaces. Data were collected for both conditions on each of the surfaces. Data were collected for both the uphill and downhill knees. Knee kinematic and kinetic data were recorded. The external knee rotation is greatest on large ballast. The knee adduction moments are greatest for the downhill knee on the hard surface sloped condition. The uphill knee has the smallest knee adduction moments. Additional models and further testing are needed to better understand joint loading while walking on crushed rock aggregate.
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INTRODUCTION

Background

Walking continues to be an important component of today’s standard of living. Knee injuries may impede or prevent independent walking; creating a physical limitation or a disability that reduces quality of life (Jackson, Wluka, Teichtahl, Morris, & Cicuttini, 2004). Osteoarthritis (OA) is a chronic condition that is characterized by the degradation of articular cartilage. It is estimated that 20% of adults are affected by OA (Bradley, Rasooly, & Webster, 1994). The knee is the weight bearing joint most commonly affected by OA (Andriacchi, Lang, Alexander, & Hurtwitz, 2000). OA may create severe functional limitations, which can lead to joint replacement surgery; it is the most common reason for total knee replacement (Corti & Rigon, 2003). OA of the knee has a prevalence of 13.9% in adults over the age of 25 (Lawrence, Felson, Helmick, Arnold, Choi, Deyo, Gabriel, Hirsch, Hochberg, Hunder, Jordan, Katz, Kremers, Wolfe, 2007) and population data suggest that the prevalence of OA increases with age and body mass index (BMI). To date, the exact mechanisms that lead to OA remain unclear.

High dynamic joint loads and repetition have been reported to have an association with disease development and severity (Andriacchi, 1994; Sharma, et al., 1998). It has been reported that muscle co-contraction may be used to help stabilize the knee joint (Schipplein & Andriacchi, 1991). While co-contraction can help stabilize and disperse the load evenly across the joint, it also contributes to an increase in total joint load. This increase in load may accelerate joint degradation. Obesity, as described by BMI, has been shown to increase the risk of OA 35% per each 5 kg body weight increase over normal (Felson, Goggins, Niu, Zhang, & Hunter, 2004).
Previous acute knee injury, previous fracture or severe twisting resulting in a sprain, has been reported as a significant risk factor of up to 7.4 times more likely to develop OA following the knee injury (Wilder, Hall, Barrett, & Lemrow, 2002).

The loads on the knee are dispersed across the lateral and medial menisci. The menisci dissipate loads and reduce friction. While walking, the loads in the knee are not evenly distributed across the menisci during the gait cycle. The medial meniscus is loaded approximately 2.5 times greater than the lateral meniscus (Baliunas, Hurwitz, Ryals, Karrar, Case, Block, Andriacchi, 2002). This asymmetric loading of the knee may explain the higher prevalence of OA in the medial compartment. Approximately 60-80% of the compressive load at the medial meniscus is produced by the external knee adduction moment (Baliunas, 2002).

The external knee adduction moment adducts the knee into a varus position and has been shown to be correlated with the severity of OA (Foroughi, Smith, & Vanwanseele, 2009; Amin, Luepongsak, McGibbon, LaValley, Krebs, & Felson, 2004; Miyazaki, Wada, Kawahara, Sato, Baba, & Shimada, 2002; Schipplein, 1991). The knee adduction moment is calculated by determining the direction of the ground reaction force. The perpendicular distance from the ground reaction force vector to the center of the knee defines the lever arm used in determining the knee adduction moment. The adduction moment has been proposed as an indirect way to measure medial knee joint loading (Amin, 2004; Miyazaki, 2002; Schipplein, 1991). Medial compartment OA can be accompanied by medial articular cartilage degeneration. This degeneration leads to decreased medial joint space, shifting the knee alignment more varus thus increasing the adduction moment, which increases the rate of degeneration (Miyazaki, 2002).

The activities that lead to the development of OA are not fully understood. The pathogenesis of the disease is likely to be multifactorial, making it difficult to design longitudinal
studies that control for all covariates. A study by Cooper, McAlindon, Coggon, Egger, and Dieppe (1994) showed that occupations involving squatting or kneeling for more than 30 minutes per day or climbing more than 10 flights of stairs per day, had a significant increase in risk for knee OA. Heavy lifting was not a significant individual risk factor, but it did appear to increase the risks when combined with kneeling or squatting. The current understanding of the causes of knee OA, specifically as related to occupational activities, is confounded by inconsistencies in the research. It has been shown that a higher knee adduction moment is an indicator of disease severity. It remains unclear if knee adduction moment leads to disease severity and malalignment or if the disease severity and malalignment lead to the adduction moment (Foroughi, 2009).

The majority of epidemiological studies on knee OA have shown that obesity and age appear to be major factors contributing to disease development (Cicuttini, Baker, & Spector, 1996; Coggon, Reading, Croft, McLaren, Barrett, & Cooper, 2001; Cooper, Snow, McAlindon, Kellingray, Stuart, Coggon, Dieppe, 2000; Eaton, 2004; Powell, Teichtahl, Wluka, & Cicuttini, 2005). Other studies have supported the claim that through weight loss the weight on the load bearing joint was reduced, resulting in some relief from OA related pain (Focht, Rajeski, Ambrosius, Katula, & Messier, 2005; Jordan, Arden, Doherty, Bannwarth, Bijlsma, Deippe, Gunther, Hauselmann, Herrero-Beaumont, Kaklamanis, Lohmander, Leeb, Lequesne, Mazieres, Martin-Mola, Pavelka, Pendleton, Punzi, Serni, Swoboda, Verbruggen, Zimmerman-Gorska, Dougados, 2005).

Defining which physical activities are related to knee OA remains a challenge. Due to the many possible contributors to OA development and progression, it has been difficult to control for all contributing factors in previous studies.
Railroad companies use crushed rock ballast in the railroad yards and along the main rail lines. Ballast is used to support the railroad tracks and provide drainage. The ballast creates an irregular walking surface for the employees. Few studies published have investigated walking on ballast. Andres et al. (2005) investigated rear foot motion while walking on ballast of different sizes and slopes. Wade and Redfern (2007) investigated using a force plate to measure ground reaction forces while walking on ballast. Merryweather (2008) investigated the lower limb biomechanics while walking on ballast for both slanted and level surfaces.

Andres et al. (2005) compared ankle kinematics for three walking surfaces. Markers were placed on the calf and on the heel of the boot. The angle was measured for changes and movement on a hard flat, small ballast, and large ballast. A significantly greater amount of rear foot motion was found for walking on large ballast. It was concluded that the increased motion would lead to increased stresses at the knee, specifically greater torsional loading on the menisci. The study had 5 subjects. Some of the weaknesses of the study were no joint data were collected at the knee, the ballast tracks were too short to allow for regular gait, and no joint kinetics were measured.

The data for this study were partially derived from a subset of data collected by Merryweather (2008). A representative trial was selected for each subject and was used for the comparisons. Merryweather found that walking on ballast resulted in a reduced walking speed, a greater knee flexion angle for the sloped condition, and a greater knee adduction moment for the upslope limb. The study was limited by a relatively small sample size and no internal joint measures of mechanical joint and soft tissue loads.
Research Purpose

This study investigates how the knee is affected by walking on different surface conditions. Kinetic and kinematic data were obtained over the different walking conditions in order to investigate what risk factors for osteoarthritis may be present.

The first hypothesis is that walking on a sloped surface will result in higher risks for osteoarthritis than normal flat surface walking. The second hypothesis is that walking on large and/or small ballast will result in higher risks for knee OA than normal hard surface walking. Third, it is hypothesized that the downhill knee will be at higher risk for OA than the uphill knee.
METHODS

Experimental Design

Surface condition (aggregate or hard), surface configuration (sloped or flat), and uphill or downhill leg are the independent variables. A hard surface, large ballast surface, and small ballast surface conditions were investigated. Surface configurations include a normal level surface (flat) and a slanted surface with a 7° slope in the transverse plane. The sloped surface represents the extreme of previously measured slopes in a railroad yard (Merryweather, 2008) and similar experimental conditions as measured by Andres (2005).

The aggregate met the specifications for Union Pacific and Burlington Northern Santa Fe Railroads for main line (large) ballast and yard (small) ballast. Ballast gradations are shown in Table 1. One track was filled with main line ballast and the other with yard ballast. Each track was filled with 15-20 cm deep with aggregate, which was slightly compacted to minimize shifting during data collection. A hard surface made from structural plywood was placed over the yard ballast track to be used for the hard surface trials. The tracks were made 76 cm wide and 7.3 m long (Figure 1).

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<td>Percent weight passing each opening</td>
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<td>Opening sizes of sieve</td>
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<tr>
<td>Main Line (Large) Ballast</td>
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<td>Yard (Small) Ballast</td>
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The tracks were placed on adjustable jacks. The same tracks could be used both for the level surface and the sloped surface trials. The force plate (model OR6-5-1000, AMTI, Watertown, MA) was embedded in the track. The small and main line ballasts cause significant dispersion of the surface forces to where the force plate was located. A custom force plate isolation fixture was developed to isolate the track aggregate from the aggregate directly over the force plate (Figure 2). The fixture was found to effectively isolate the force plate and allow for the force plate to accurately measure applied forces (Merryweather, 2008).

Data for this study were collected as part of a previous study (Merryweather, 2008). Merryweather did not use the entire collected data set. Rather he used 20% of the collected trials. This study included the complete data set for evaluation. Motion data were collected at 60 Hz using a five camera Vicon Motus Video acquisition system (Vicon Motion Systems, Lake...
Figure 2  Force plate isolation fixture
Panasonic GS55 video cameras were used to capture the video. The analog force plate data were recorded at 600 Hz. The global coordinate system was used for all trials, maintaining the directions of the reaction force components in any walking plane slope angle. The right hand coordinate system with the positive x-axis in the direction of motion, the positive y-axis right-to-left, and the positive z-axis upward.

The study was approved by the University of Utah's Institutional Review Board (IRB). Participants were brought to the Ergonomics and Safety laboratory where they were interviewed to ensure they met all enrollment requirements. Basic anthropometric data were measured and the participants were outfitted with reflective markers. Marker placement was done according to a modified Helen Hayes Hospital Marker Set (Kadaba, Ramakrishnan, & Wootten, 1990). Each participant was given a new pair of model 2408 Red Wing work boots. The markers on the foot and ankle were placed on the boots bilaterally over the second metatarsal, heel, and lateral malleolus.

Participants were asked to wear dark colored shorts and a dark colored shirt with the sleeves and waist removed. This exposed the landmarks for placing the markers on the waist and prevented the clothing from covering any of the markers during movement.

A fourth order zero-lag digital Butterworth filter with a cutoff frequency of 6 Hz was used to condition the raw marker position data. The position data were also smoothed using a quintic spline interpolation algorithm (Woltring, 1986).

The marker positions provide unique points in space that are used to identify the location of body segments for each point in time. The locations of the limbs are used to calculate joint positions, velocities, and accelerations using Vicon Motus Video Gait Template equations (Vaughan, Davis, & O'Conner, 1992). Calibration was done with the computer based motion analysis program Motus (ViconPeak, CO, USA). The calibration was repeated for each
track condition prior to the data collection. Coordinate systems are established for each lower extremity segment, providing clinically meaningful values for comparison to published data. Joint and segment reaction forces and moments were determined using an inverse dynamics approach.

**Data Collection**

The experimental conditions (surface and condition) were randomized. Participants were allowed to take as many practice trials as they needed to feel comfortable walking on each surface prior to data collection. Participants needed to find their preferred starting location on the track in order to land the appropriate foot on the force plate. Trials where the participant either missed or partially missed landing the entire foot on the force plate were discarded and the participant was asked to repeat that trial. Ten good trials were collected for each experimental condition. The 10 trials consisted of 5 trials where the uphill foot contacted the force plate and 5 for the downhill foot.

Upon completion of 10 acceptable trials, the participant was asked to step down from the track. The participant was given a rest time while the researchers setup the next randomly selected walkway. The participant would then do 5 more trials for each foot; this was repeated until 10 trials were collected for each condition. The walking direction was in the same direction for all trials, so the right foot was always on the uphill side of the slope and the left foot was always on the downhill side of the slope. Each participant completed at least 60 trials (5 trials x 3 surfaces x 2 conditions x 2 feet).

Upon completion of the 60 trials, the markers were removed and the participants were asked to fill out a questionnaire about how the experimental conditions of the study compared to their walking environment at work. An average of approximately 4 hours per session was
needed for each participant, ninety minutes for the walking portion and 120 minutes for setup and calibration.

**Data**

Two full gait cycles were digitized for each trial to ensure sufficient data were recorded. The heel strike when stepping onto the force plate was always taken as the initial event. The following heel strike of the same foot was used as the final event for one complete cycle.

The trials had all been digitized as part of a previous study (Merryweather, 2008). However, some of the unused trials were reprocessed to improve reliability and reduce errors. Though the markers were digitized, the majority of them had not been used in the results reported by Merryweather. The desired output from Peak Motus differed from what was done previously, so each trial required new calculation programming. Approximately half of the trials needed to have gait events manually selected through observation of the video and input into the system. Event detection was accomplished by visually determining when the subject accomplished heel-strike or toe-off.

Kinetic and kinematic data were obtained for the knee only. Recorded knee data include internal/external angle, flexion/extension angle, abduction/adduction angle, anterior/posterior force, medial/lateral force, proximal/distal force, resultant force, internal/external moment, flexion/extension moment, abduction/adduction moment, and resultant moment. Forces were normalized to bodyweight (N/N). Moments were normalized to bodyweight x height (Nm/Nm).

In order to compare data, each stride was normalized to 101 data points for one gait cycle. Reliability of the data collection was measured by taking the Pearson product moment correlation coefficient in a test retest measure. The five trials for each condition were tested for
reliability for each subject. Any trial not correlating ($r_{tt} < 0.55$) with the others was inspected for errors. In some trials the participants stumbled or had an abnormal gait compared with other trials; abnormal trials were removed from the data set. The remaining trials were averaged, resulting in one representative data set per subject per condition.

The averaged trials were then imported into the statistical software (SAS v9.3). Tukey’s test was run in conjunction with a general linear model (GLM) regression analysis. The GLM investigated the differences between the following variables: condition (flat/sloped), surface (hard, large ballast, small ballast), and foot (right/left).

Participant Requirements

Participants all met the following requirements:

- Age: 18-60
- BMI: Preferably between 18.5-24.9
- Normal Gait Patterns
- No abnormal foot physical features
  - Club Feet
  - Flat Feet
  - Extreme Valgus or Varus
  - Foot problems that interfere with walking (such as a callus, corn, ingrown toenail, wart, pain, skin ulcer, swelling, spasms).

Ten railroad workers from Salt Lake City, Utah were selected to represent a healthy population of railroad workers. A sample size of 10 is twice as large as other similar studies (Andres, Holt, & Kubo, 2005; Wade & Redfern, 2007). Participants had at least 3 years experience walking on ballast in the job setting. The participants consisted of conductors,
switchmen, and other workers employed in positions involving walking on ballast in a train yard on a regular basis. Prior to participation each participant read and signed an informed consent form approved by the research IRB. The average participant was overweight as defined by BMI (Table 2).

<table>
<thead>
<tr>
<th>Age (SD)</th>
<th>Years with Railroad (SD)</th>
<th>Height-m (SD)</th>
<th>Weight-Kg (SD)</th>
<th>BMI (SD)</th>
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<td>37.1 (8.94)</td>
<td>8.4 (7.7)</td>
<td>1.8 (0.08)</td>
<td>84.0 (14.14)</td>
<td>27.4 (4.3)</td>
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RESULTS

After performing the reliability analysis the average correlation of trials by subject was 0.93 ± 0.04. 131 trials were removed due to an irregular gait. An average of 3.91 ± 1.17 trials were used per subject per condition from the five recorded trials.

At heel-strike the abduction/adduction (ABD/ADD) angles were near zero. The knee then became abducted to approximately 5° until toe-off, where it again approached zero degrees. The swing phase resulted in abduction angles up to 17° at the peak and then returned to zero. The knee remained abducted for most of the stride under all conditions (Figures 3, 4 and 5).

There were only minor differences in ABD/ADD knee angles between flat and sloped conditions (p=0.226), most of which appeared during the swing phase of the stride. The ABD/ADD knee angles were significantly different between hard and large ballast (p=0.032) and hard and small ballast (p=0.024) ballast surfaces. There was no significant difference in ABD/ADD knee angles between small and large ballast (p=0.918). Left knee ABD/ADD angles were significantly different between flat and sloped conditions (p=0.003). None of the ABD/ADD knee angle data was significantly different for the uphill knee. The ABD/ADD knee angles were significantly different between the uphill and downhill knees (p<0.0001).

The flexion/extension (FLX/EXT) knee angles all followed the same trend as expected from literature. At heel-strike the angle was near zero degrees. There was a small flexion of 10-20° as pressure increases on the heel. When the pressure changed from the heel to the toe the
Figure 3 Left knee abduction/adduction (ABD/ADD) angles. Adduction is positive and abduction is negative.
Figure 4 Right knee ABD/ADD angles. Adduction is positive and abduction is negative.
Figure 5 ABD/ADD angles on different surfaces. Adduction is positive and abduction is negative.
flexion angle again approached zero. Prior to the toe-off event the flexion angle began increasing until it peaked at approximately 60-70°. The angle then returned to zero degrees prior to the next heel-strike (Figures 6 and 7).

The left knee FLX/EXT angles were significantly different between surfaces (p=0.002). The FLX/EXT angles were significantly different between sloped and flat conditions for the right knee (p<0.0001). The FLX/EXT knee angles were significantly different between uphill and downhill knees (p<0.0001).

The internal/external (INT/EXT) knee angles followed a similar trend for all trials. For most of the stride the knee remained internally rotated. As the toe is leaving the ground the knee becomes externally rotated (Figures 8 and 9).

The INT/EXT knee angles were significantly different between surfaces (p<0.0001). There were no INT/EXT knee angle differences between conditions (p=0.388). The INT/EXT knee angles were significantly different between uphill and downhill knees (p<0.0001).

Medial/lateral (MED/LAT) knee forces had an initial lateral force at heel-strike, followed by a greater medial force during the loaded portion of the stride. The greatest medial forces reached approximately 15% BW. The MED/LAT forces returned to zero as the foot pushed off prior to the toe-off event (Figures 10, 11 and 12).

The left knee MED/LAT forces were significantly different between surfaces (p<0.0001) and between conditions (p<0.0001). The right knee MED/LAT forces were not significantly different between surfaces (p=0.105) nor between conditions (p=0.354). The MED/LAT knee forces were significantly different between uphill and downhill knees (p<0.0001).

The anterior/posterior (ANT/POS) knee forces remained anterior during the majority of the stance phase. There is an initial posterior force immediately following heel-strike. The force
Figure 6 Left knee FLX/EXT angles. Flexion is positive.
Figure 7 Right knee FLX/EXT angles. Flexion is positive.
Figure 8  INT/EXT angles on different surfaces. External is positive and internal is negative.
Figure 9 INT/EXT angles for uphill and downhill knees. External is positive and internal is negative.
Figure 10  Left knee MED/LAT forces. Lateral is positive and medial is negative.
Figure 11 Right knee MED/LAT forces. Lateral is positive and medial is negative.
Figure 12  MED/LAT forces on different surfaces. Lateral is positive and medial is negative.
then becomes posterior for an initial peak of 10-20% BW followed by a larger (40% BW) peak force prior to toe-off (Figure 13).

ANT/POS knee forces were not significantly different between conditions ($p=0.448$). The left knee ANT/POS forces were significantly different between surfaces ($p=0.0004$). The right knee ANT/POS forces were not significantly different between surfaces ($p=0.336$). The ANT/POS knee forces were significantly different between uphill and downhill knees ($p=0.0009$).

The proximal/distal (PRX/DIS) knee forces were approximately zero during the swing phase. At heel-strike a large (>100% BW) proximal force was loaded. This decreased after the initial spike until the toe-off loading occurred. The distal force again increased to approximately 100% BW. The distal force then decreases and a small (<0.1% BW) proximal force occurred during the swing phase (Figure 14).

The PRX/DIS knee forces were not significantly different for any of the measurements. The hard and small ballast have higher peak proximal forces than the hard surface.

The ABD/ADD knee moments were approximately zero during the swing phase. After the initial heel-strike the knee experienced the peak adduction moment. The moment then decreased until the loading for the push-off. There was a lesser increase and peak in the adduction moment before it returned to zero for the swing phase (Figures 15, 16, 17 and 18).

The ABD/ADD knee moments were significantly different between surfaces ($p<0.0001$). The ABD/ADD knee moments were also significantly different between conditions ($p<0.0001$) and between uphill and downhill knees ($p<0.0001$).

There was an initial flexion moment following heel-strike, which turned to the peak extension moment as the knee is loaded. The moment turned from an extension during heel-strike and initial loading to a flexion moment during the push-off with the toe, prior to the toe-
**Figure 13** ANT/POS forces on different surfaces. Posterior is positive and anterior is negative.
Figure 14 PRX/DIS forces on different surfaces. Distal is positive and proximal is negative.
Figure 15  Left knee ABD/ADD moments.  Adduction is positive and abduction is negative.
Figure 16 Right knee ABD/ADD moments. Adduction is positive and abduction is negative.
Figure 17 ABD/ADD moments on different surfaces. Adduction is positive and abduction is negative.
Figure 18 ABD/ADD moments for the uphill and downhill knees. Adduction is positive and abduction is negative.
off. The moment returned to near zero after toe-off before experiencing small flexion moment during the swing phase (Figures 19, 20, 21 and 22).

The left knee FLX/EXT moments were significantly different between surfaces ($p<0.0001$). The right knee FLX/EXT moments were significantly different between surfaces ($p=0.0004$) and between conditions ($p<0.0001$). The FLX/EXT knee moments were significantly different between uphill and downhill knees ($p<0.0001$).

The internal/external (INT/EXT) knee moments had an initial exterior moment at heel-strike. As the weight was transferred from the heel to the rest of the foot the moment changed to an internal moment. There was a peak internal moment at approximately 50% of the stride. The moment then returned to zero for the swing phase of the stride (Figures 23, 24, 25, and 26).

INT/EXT knee moments were significantly different between surfaces ($p<0.0001$). The INT/EXT knee moments were significantly different between conditions ($p=0.03$). The INT/EXT knee moments were significantly different between uphill and downhill knees ($p<0.0001$).
Figure 19 Left knee FLX/EXT moments. Flexion is positive and extension is negative.
Figure 20  Right knee FLX/EXT moments. Flexion is positive and extension is negative
Figure 21 FLX/EXT knee moments on different surfaces. Flexion is positive and extension is negative.
Figure 22 FLX/EXT moments for the uphill and downhill knees. Flexion is positive and extension is negative.
Figure 23 Left knee INT/EXT moments. External is positive and internal is negative.
Figure 24 Right knee INT/EXT moments. Exterior is positive and internal is negative.
Figure 25  INT/EXT knee moments on different surfaces. External is positive and internal is negative.
Figure 26 INT/EXT moments for the uphill and downhill knees. External is positive and internal is negative.
DISCUSSION

The goal of this study is to better understand the moments at the knee as a function of walking surface and condition and how it relates to the risk for knee OA. The knee moments were investigated and compared to what current literature has shown as indicators of knee OA progression and development (Amin, 2004; Andriacchi, 2000; Baliunas, 2002; Foroughi, 2009; Miyazaki, 2002; Mündermann, Dyrby, & Andriacchi, 2005; Sharma, Hurwitz, Thonar, Sum, Lenz, Dunlop, Schnitzer, Kirwan-Mellis, & Andriacchi, 1998;).

Knee Angles

The knee ABD/ADD angles measure large differences ($p<0.0001$) when comparing right to left knees. This large difference could be explained by the marker placement. The markers could have been set up differently for the right from the left leg. The right and left knees are expected to have matching kinematics while walking on a flat surface for normal gait.

The swing phase shows some differences in the ABD/ADD angles. The stance phase is of a greater interest, however, due to the effects on the knee being greater while the knee is loaded. The most notable differences are for the downhill knee on the hard surface, where the sloped condition has a higher adduction angle ($p=0.003$). This increased adduction angle increases the moment arm for the adduction moment and may be contributing to increased medial meniscus loading. The large and small ballast surfaces do not have significant differences between conditions. The lack of any differences on the aggregate may be attributed to an
increased co-contraction to help stabilize the knee. The co-contraction may hold the joint tight, not allowing for an increased adduction angle.

The greater flexion angles on the sloped trials may be due to the need for greater flexion of the right knee when it is on the uphill side of the slope. While in the stance phase of the stride more work would be required for the right knee to flex at the same angle on the sloped condition as it does on the flat condition. The entire body would need to be elevated to the level of the uphill side of the track. It is also necessary to flex the right knee more during the swing phase in order to prevent colliding with the ground.

The INT/EXT angle differences between right and left knee may be due to marker placement. There are no measurable differences in conditions. The differences in the INT/EXT knee angles on different surfaces may be attributed to the need for increased stability on the loose ballast. The ankle is rotated externally to increase stability. The large ballast has the most externally rotated angles for the right knee, followed by the small ballast. The hard flat surface has the least external rotation. The downhill knee INT/EXT angles were more externally rotated than the uphill.

**Knee Forces**

The left knee has the greatest medial forces on the hard surface, the large ballast has the next highest and the small ballast has the least. This may be because of the slower walking velocity on the large and small ballast. The right knee has greater medial forces than the left ($p<0.0001$). The right knee does not have significantly higher medial forces for the sloped condition ($p=0.354$) when comparing the means; however it is higher during the stance phase by 15%. The left knee has higher medial forces for the sloped condition ($p=0.0001$).
The greater medial forces on the right knee can be attributed to the direction of the ground reaction force and its perpendicular distance from the center of the knee. On a sloped surface the center of mass will shift toward the lower limb. The cause of the increased medial force on the left knee while on the sloped configuration is unclear.

MED/LAT forces may trigger co-contraction to aid in stabilizing the joint. Co-contraction may increase the load on the joint, increasing the rate of degeneration (Schipplein & Andriacchi, 1991).

ANT/POS knee forces are different on the different surfaces. The left knee has the most notable differences between the small ballast and the other surfaces. The small ballast has larger posterior forces throughout the stance phase. There are significant differences between right and left knee forces (p=0.012).

There are no significant differences between PRX/DIS knee forces for the different surface conditions. The large and small ballast surfaces have larger peak proximal forces than the hard surface. The increased proximal forces, for the large and small ballast surfaces, occur during the initial peak force. The force required for toe-off did not have large differences in surfaces. It was expected to have a larger initial proximal force on the hard surface. It is unclear why the initial proximal force is greater on aggregate surfaces.

Knee Moments

The adduction moments for the left knee are statistically significantly greater for the sloped condition than for the flat condition (p<0.0001). This is partly due to the shear force from the slope. The shear force translates into an internal lateral force at the left knee. The added internal lateral force necessitates a larger external adduction moment.
The hard ballast surface has a large difference in the ABD/ADD moments between conditions. The large and small ballast surfaces have smaller differences, though still significant. The sloped condition on the hard surface had the highest adduction moment, the average is approximately 70% larger than the others.

The adduction moments for the right knee are statistically significantly smaller for the sloped condition than for the flat condition ($p<0.0001$). This is partially attributed to the shear force from the slope. For the right knee the shear force works opposite of how it is created for the left knee. The internal lateral force is decreased and there is less need for an adduction moment.

The hard surface has the highest peak adduction moments for the right knee. The differences are not as large for the right knee as they are for the left. The small and large ballast have higher adduction moments, than the hard surface, during the push leading to the toe-off event of the stance phase for the sloped condition. The hard surface was more affected by the condition than the small and large ballast surfaces.

The left knee extension moments are larger on the sloped condition during the initial loading of the joint and larger on the flat condition flexion moments during the push before the toe-off. The difference is most notable in the small ballast trials. The right knee has higher peak FLX/EXT moments on the sloped surface for both the initial extension and the flexion prior to toe-off. When the weight is shifted from the heel to the toe, the uphill has a larger difference in condition than the downhill knee. The small and large ballast can shift under the foot, this affects the FLX/EXT moments prior to toe-off. There is a greater extension moment prior to the toe-off on the ballast surfaces than the norm, the small ballast has the greatest moment.
The duration of the loaded portion of the FLX/EXT moment is longest for the hard surface, followed by the large ballast and then the small ballast has the shortest duration. This lengthens the time of the double support.

The left knee internal moments are greater for the sloped condition than the flat (\(p<0.0001\)). This difference is most notable on the hard surface. Increased internal moment while on the slope counters the increased shear force due to the slope. The sloped condition moments are higher after the center of pressure leaves the heel and forms a moment arm to the center of the knee.

The internal knee moment is highest for the hard surface, then the large ballast, and the small ballast has the lowest moment (\(p<0.0001\)). Both the right and left knees follow the same trends when comparing the differences made by the surfaces. On the hard surface the left knee has a higher internal moment than the right.

The internal moments are higher for the sloped condition than the flat (\(p<0.030\)). This is most noticeable on the hard surface. The large and small ballast surfaces also have higher internal moments on the sloped surface, yet the peak internal moments are approximately the same for both conditions.

Walking on the sloped surface increases the internal moments for the left knee and decreases for the right, due to the direction of the shear force on the slope. The shear force pushes the toe of the left foot in a medial direction and pushes the right toe laterally. This resultant moment is in an internal direction at the left knee and external at the right.
CONCLUSION

Study Findings

The knee ABD/ADD moments have been shown to be an accurate measurement of medial meniscus loading (Amin, 2004; Foroughi, 2009; Miyazaki, 2002; Schipplein, 1991). Greater loading of the medial meniscus is a risk for accelerated cartilage degradation and knee OA.

Table 3 is a list of the means and standard deviations for the data of interest. Table 4 lists some of the significant findings of this research. The downhill knee has higher adduction moments when walking on sloped surfaces ($p<0.0001$). The uphill knee has smaller adduction moments on the sloped surface ($p<0.0001$). Walking on sloped surfaces may contribute to the development knee OA on the downhill knee.

The exterior knee rotation is greatest on the large ballast surface. The small ballast surface also has a higher external angle than expected. External rotation has been shown to reduce the adduction moment and relieve loading of the medial meniscus (Jenkyn, 2008). This is one mechanism that accounts for the reduced adduction moment on small and large ballast surfaces.

The first hypothesis that walking on a sloped surface will result in higher risk for knee OA is supported for the downhill knee. The uphill knee has less medial meniscus loading while on a sloped surface and would then be at lower risk for knee OA. A worker may rotate which knee is the downhill knee while at work, but this is not yet understood if applying a greater load
Table 3 Summary of Means With Their Standard Deviations

<table>
<thead>
<tr>
<th></th>
<th>Left Knee</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Hard</td>
<td>Large Ballast</td>
<td>Small Ballast</td>
<td>Flat</td>
<td>Slope</td>
</tr>
<tr>
<td>ABD/ADD Angles</td>
<td>-5.68(4.98)</td>
<td>-6.11(4.97)</td>
<td>-5.91(5.06)</td>
<td>-6.17(5.01)</td>
<td>-5.63(5.0)</td>
</tr>
<tr>
<td>INT/EXT Angles</td>
<td>-8.86(12.68)</td>
<td>-2.81(10.11)</td>
<td>-7.89(13.39)</td>
<td>-6.60(12.12)</td>
<td>-6.44(12.0)</td>
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<tr>
<td>MED/LAT Forces</td>
<td>-3.62(4.49)</td>
<td>-3.09(3.44)</td>
<td>-1.36(4.28)</td>
<td>-2.28(3.76)</td>
<td>-3.10(4.38)</td>
</tr>
<tr>
<td>ABD/ADD Moments</td>
<td>1.58(0.61)</td>
<td>0.97(0.51)</td>
<td>1.65(0.57)</td>
<td>0.99(0.53)</td>
<td>1.41(0.60)</td>
</tr>
<tr>
<td>FLX/EXT Moments</td>
<td>-0.29(0.76)</td>
<td>0.07(0.87)</td>
<td>-0.40(0.96)</td>
<td>-0.20(0.75)</td>
<td>-0.20(0.97)</td>
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</table>

<table>
<thead>
<tr>
<th></th>
<th>Right Knee</th>
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<th></th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Hard</td>
<td>Large Ballast</td>
<td>Small Ballast</td>
<td>Flat</td>
<td>Slope</td>
</tr>
<tr>
<td>ABD/ADD Angles</td>
<td>-4.68(6.23)</td>
<td>-4.99(6.76)</td>
<td>-5.23(6.89)</td>
<td>-4.87(6.53)</td>
<td>-5.07(6.72)</td>
</tr>
<tr>
<td>INT/EXT Angles</td>
<td>-5.58(9.22)</td>
<td>-1.97(11.19)</td>
<td>-2.90(10.15)</td>
<td>-3.22(10.03)</td>
<td>-3.75(10.34)</td>
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<tr>
<td>MED/LAT Forces</td>
<td>-3.77(3.52)</td>
<td>-3.59(3.78)</td>
<td>-4.07(4.63)</td>
<td>-3.73(4.18)</td>
<td>-3.88(3.78)</td>
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<tr>
<td>ABD/ADD Moments</td>
<td>0.70(0.54)</td>
<td>0.95(0.53)</td>
<td>0.87(0.72)</td>
<td>1.05(0.63)</td>
<td>0.63(0.56)</td>
</tr>
<tr>
<td>FLX/EXT Moments</td>
<td>-0.41(0.75)</td>
<td>-0.19(1.06)</td>
<td>-0.34(1.09)</td>
<td>-0.11(0.96)</td>
<td>-0.52(0.97)</td>
</tr>
</tbody>
</table>

Table 4 Summary of Significant Findings

<table>
<thead>
<tr>
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<td>p values</td>
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<td>Right Knee</td>
<td></td>
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<tr>
<td>Surface Condition</td>
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<td>Small Ballast</td>
<td>Flat</td>
<td>Slope</td>
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<td>ABD/ADD Angles</td>
<td>0.056</td>
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<td>0.357</td>
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<tr>
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<td>&lt;0.0001</td>
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<td>&lt;0.0001</td>
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</tr>
<tr>
<td>MED/LAT Forces</td>
<td>0.007</td>
<td>&lt;0.0001</td>
<td>&lt;0.0001</td>
<td>&lt;0.0001</td>
<td>0.013</td>
</tr>
<tr>
<td>ABD/ADD Moments</td>
<td>&lt;0.0001</td>
<td>&lt;0.0001</td>
<td>0.111</td>
<td>&lt;0.0001</td>
<td>&lt;0.0001</td>
</tr>
<tr>
<td>FLX/EXT Moments</td>
<td>&lt;0.0001</td>
<td>&lt;0.0001</td>
<td>0.896</td>
<td>&lt;0.0001</td>
<td>0.236</td>
</tr>
</tbody>
</table>

The second hypothesis that walking on large or small ballast would result in higher risks for knee OA is not supported by this research. Walking on ballast does not increase the knee adduction moment. The decreased magnitude of the adduction moment on the small and large ballast may be partially attributed to the increased external rotation. The adduction moment is also affected by walking velocity. Merryweather (2008) reported that walking velocity was
reduced with the size of aggregate. The actual loading at the knee joint, however, may also be
affected by muscle co-contraction.

The third hypothesis that the downhill knee is at greater risk for knee OA than the uphill
knee is supported by the results. The downhill knee is subject to an increased adduction
moment which translates into an increased load on the medial meniscus with the possibility of
an accelerated degradation of the cartilage.

**Suggestions for Future Work**

One limitation of this study is that all forces and moments were calculated from external
forces through reverse dynamics. Future work should look at muscle activation to determine if
co-contraction is present on the unstable ballast surfaces. Future work should also include the
use of software designed for internal loads including muscles, ligaments and bones.

This study looked specifically at the knee. Future work should look more
comprehensively at the entire body. Upper body biomechanics along with the other joints of
the lower limbs may improve the understanding of what contributes to loading of the knee
joint.
REFERENCES


