# Joint loading, muscle co-contraction, ligament force and peak knee contact forces when walking on railroad ballast

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**Abstract:** Knee contact force (KCF), muscle co-contraction and ligament forces at the knee were determined for eight railroad workers walking on ballast. Independent variables included: surface condition [no ballast (NB), walking ballast (WB), mainline ballast (MB)], configuration (level, 7° lateral slant), and uphill/downhill limb. KCF was not affected by surface condition or surface configuration. Muscle co-contraction was higher for WB than NB, and higher in the uphill than downhill limb. First peak KCF in the anterior cruciate and medial collateral ligaments were higher and second peak KCF in the lateral collateral ligament were higher in the lateral slant condition than the level condition. Force in the medial collateral ligament was higher for the uphill than for downhill limb. Force in the anterior cruciate and lateral collateral ligaments was higher for downhill than uphill. This suggests that railroad worker gait may include compensatory mechanisms to reduce peak KCFs and knee instability.

**Keywords:** railroad ballast; knee contact force; KCF; ligament force; muscle co-contraction.

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## 1 Introduction

While there has been some research investigating human gait on irregular surfaces in occupational environments, such as railroad ballast, little research has focused on the peak knee contact force (KCF) and biomechanical forces acting at the knee when walking on ballast. The goal of this study was to better understand the biomechanical loadings at the knee during peak KCF for different surface conditions and configurations when walking on ballast.

Railroad workers are required to walk and perform various tasks on crushed rock aggregate (ballast). The two ballast types are defined as walking ballast (WB, 9.53 mm to 31.75 mm), generally located in railroad yards, and mainline ballast (MB, 19.05 mm to 63.5 mm) generally located on main track lines (Andres et al., 2005; Wade et al., 2010; Merryweather, 2008). The smaller walking ballast is generally used in rail yards and other locations where workers must walk and perform inspection/maintenance activities. The larger mainline ballast is generally used along the main tracks between stations where train velocity is higher and track stability and drainage is most important. According to the Federal Railroad Administration (FRA), walking injures are common and contributed 15.2% to 16.5% of all injuries and accounted for 18.6% to 19.6% of days absent from work between 2005 and 2009 (FRA, 2005–2009). The actual physical effect on railroad workers from walking on ballast and the etiology of these injuries is unclear. Previous research has focused on the kinetic and kinematic characteristics of walking on ballast. Andres et al. (2005) investigated rear foot motion when walking on ballast with five healthy male subjects and found that walking on MB significantly increased rear foot range of motion compared to walking on either WB or no ballast (NB). He proposed that this may increase stresses applied to the knee joint since rear foot eversion is coupled with medial rotation of the tibia. A study performed by Merryweather (2008) focused on lower limb biomechanics when walking on ballast with ten railroad workers. The main findings were that mediolateral kinetics were significantly different between level surface and laterally slanted surface, and the knee joint had a greater peak flexion when walking on ballast than NB. A follow-up study (Quincy, 2010) further reported that the knee adduction moment was larger for the downhill limb than the uphill limb throughout the gait cycle when walking on a laterally slanted surface. A recent study (Wade et al., 2010) examined the impact of ballast on gait biomechanics with 20 healthy adult males. Their findings were that walking on ballast increased muscle co-contraction levels compared with NB, and that the range of joint moments were smaller for MB and WB compared with NB. The changes in normal joint loading when walking on ballast may damage soft tissue such as the knee ligament and menisci or result in osteoarthritis (Shelburne et al., 2006). It is therefore proposed that a better understanding of biomechanical forces during peak knee joint loading when walking on ballast is important to identify risk factors for lower extremity injuries.

The purpose of this study was to investigate joint loading, muscle co-contraction and ligament force at the knee at the times of peak KCF. Specific hypotheses include:

1 peak KCF is larger during walking on ballast compared with NB

2 knee muscle co-contraction is higher when walking on ballast compared with NB during peak KCF

3 forces in the medial and lateral collateral ligaments are higher for uphill than for downhill limbs during peak KCF when walking on the lateral slant surface.

## 2 Subjects, methods, and analysis

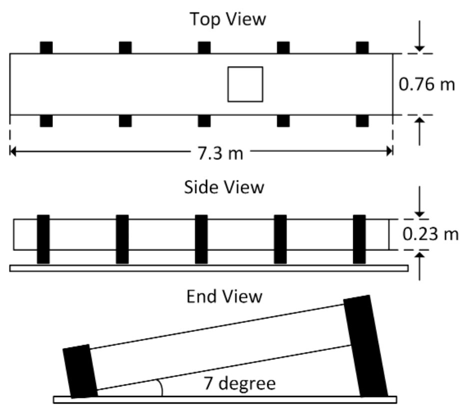
### *2.1 Subjects*

Eight male railroad workers from a local railroad facility were recruited for the study (age 39.17 ± 8.80 years old, height 1.76 ± 0.09 m, body mass 82.71 ± 14.14 kg, years with railroad 9.79 ± 8.30). Each participant signed an informed consent form approved by the University of Utah's institutional review board and received a new pair of Red Wing, model 2408 work boots for use in the study (8" ankle, steel toe, oil-resistant sole). Participant inclusion criteria included at least three years as a railroad worker, normal gait pattern and no abnormal foot physical features.

### *2.2 Methods*

Independent variables controlled for this research included surface condition (MB, WB and NB), surface configurations (level surface and 7° lateral slant surface), and uphill or downhill limb. The 7° lateral slanted surface (slanted right-to-left in this study) was selected to represent a common lateral slope along the rails in railroad yards. Previously, collected experimental kinematic and ground reaction force data were used (Merryweather, 2008). Two walking tracks (0.76 m wide by 7.3 m long) were fabricated and filled with 15–20 cm deep MB and WB, respectively (Figure 1). A hard surface made from structural plywood was placed over the WB track to be used for NB condition. These two tracks were placed on the adjustable jacks so one side of each track could be elevated to generate the lateral slant condition for each surface condition. One force plate (model OR6-5-1000, AMTI, Watertown, MA) was embedded in each track with a force plate isolation fixture, which isolated the surface force applied to the aggregate to the force plate (Figure 1).

**Figure 1** Schematic diagram of the track, the track was 0.76 m wide and 7.3 m long with the adjustable jacks on both sides




The combinations of surface conditions and configurations were randomised. For each combination, the participant was requested to perform five trials with clean force plate strikes for each limb at a self-selected speed. The walking direction was the same for all trials to keep the right limb in the uphill for the lateral slant configuration. Motion data were collected at 60 Hz with Vicon Motus Video system (Vicon Motion Systems, Lake Forest, CA) using a modified Helen Hayes marker set and conditioned using a fourth order zero-lag Butterworth filter with a cutoff frequency of 6 Hz. The 60 Hz video system was selected because it is portable and can also be used for field data acquisition. The movement patterns were relatively slow and it was observed that the 60 Hz motion data were adequate for the subsequent analysis. Ground reaction force data were collected at 600 Hz and low-pass filtered at 20 Hz.

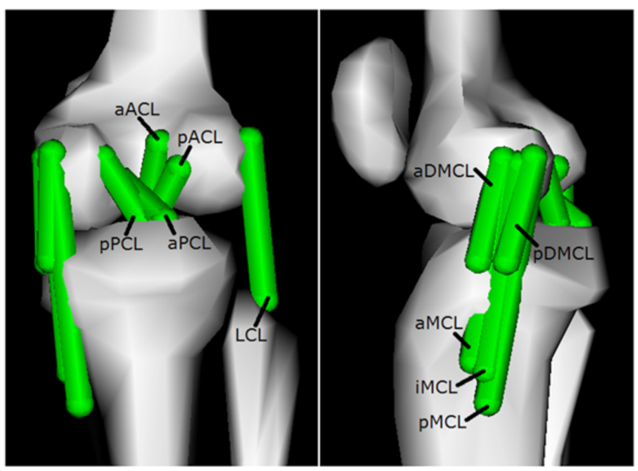
### *2.3 Data processing*

Major gait events (heel strike and toe off) were defined via force plate activation with a 20 N threshold to identify the stance phase of gait cycle. A representative trial was chosen from each group of trials for each condition/configuration combination to help reduce any effect of walking speed on the data. The criteria of the representative trial was that the walking speed was within 5% of the average speed of the participant for all trials of the same condition/configuration. Therefore, a total of 96 trials (8 subjects * 3 surface conditions * 2 surface configurations * 2 feet [uphill and downhill]) were included in the present study.

### *2.3.1 Gait model*

The musculoskeletal model developed for the study consisted of 14 rigid segments, 27 degrees of freedom (DOFs), 56 muscles and ten ligament bundles of the knee (Xu, 2013). The knee was represented as a three DOFs joint including three independent rotations with proximodistal and anteroposterior translations occurring as a function of knee flexion (Yamaguchi and Zajac, 1989). An eight-muscle system was defined in the knee joint of the model. This included five knee flexors: biceps femoris long head (BFLH), biceps femoris short head (BFSH), gracilis (GRAC), gastrocnemius (GAS), sartorius (SAR), and three knee extensors: rectus femoris (RF), vastus (VAS) and patella tendon (PT). Knee cruciate and collateral ligaments (Figure 2) were represented by ten ligament bundles. The attachment sites of ligament bundles were based on the dataset reported by Blankevoort et al. (1991). The path of ligament bundle was considered as a straight line and the effect of ligament-bone contact was neglected. Each ligament bundle was assumed as elastic and its properties were described by a non-linear, force-length curve (Blankevoort and Huiskes, 1991). The stiffness and reference strain of the ligament bundles were obtained from Pandy et al. (1998). The geometrical and mechanical properties of the ligaments were validated to approximate the behaviour of the physical knee with independent knee rotations in three body planes.

**Figure 2** The attachment sites of knee ligament bundles (see online version for colours)




Notes: The abbreviation of ligament bundles are: aACL, anterior bundle of the anterior cruciate ligament (ACL); pACL, posterior bundle of the ACL; aPCL, anterior bundle of the posterior cruciate ligament (PCL); pPCL, posterior bundle of the PCL; aMCL, anterior bundle of the superficial layer of the medial collateral ligament (MCL); iMCL, inferior bundle of the superficial layer of the MCL; pMCL, posterior bundle of the superficial layer of the MCL; aDMCL, anterior bundle of the deep layer of the MCL, pDMCL, posterior bundle of the deep layer of the MCL.

### 2.3.2 Gait simulation

OpenSim (Delp et al., 2007) was used to simulate the gait trials for each condition (surface conditions × surface configurations × uphill and downhill feet). The model was first scaled to match each subject. Segment lengths were scaled based on relative distances between pairs of markers obtained from the motion capture system and the corresponding virtual marker located in the model. The segment masses were scaled by the subject's body mass and anthropometric data. The length of the muscles and ligament bundles were also scaled to keep the attachment sites relatively fixed on the body segments. Inverse kinematics were employed to determine joint kinematics by positioning the model as a 'best match' pose, which used a weighted least squares approach to minimises both marker and coordinate errors between the motion capture data and the OpenSim model. Finally, the residual reduction and computed muscle control algorithms were used to refine the model kinematics and generate muscle excitations and forces to drive the dynamic simulations during gait.

### 2.3.2 Knee biomechanics

The joint reaction program in OpenSim was employed to calculate KCF, which represented the internal loads between the tibia and femoral cartilages and the sum of force acting on both medial and lateral compartments of the meniscus. This overall KCF was calculated as the vector sum of the knee reaction force (KRF), the compressive forces from muscles crossing the knee joint, and knee collateral and cruciate ligaments [equation (1) to (3)].

$$\vec{F}_{KCF} = \sum\left(\vec{F}_{KRF} + \vec{F}_{Muscle} + \vec{F}_{Ligament}\right) \tag{1}$$

$$\vec{F}_{muscle} = \sum\left(\vec{F}_{BFLH} + \vec{F}_{BFSH} + \vec{F}_{GRAC} + \vec{F}_{GAS} + \vec{F}_{SAR} + \vec{F}_{RF} + \vec{F}_{VAS} + \vec{F}_{PT}\right) \tag{2}$$

$$\vec{F}_{Ligament} = \sum\left(\vec{F}_{ACL} + \vec{F}_{PCL} + \vec{F}_{MCL} + \vec{F}_{LCL}\right) \tag{3}$$

Muscle co-contraction, expressed as co-contraction index (CCI), was used to describe the simultaneous activity of various muscles crossing the knee [equation (4)] (Karakostas et al., 2003).

$$CCI = \frac{\sum F_{Total}^{M}}{\sum F_{Agonists}^{M}} - 1 \tag{4}$$

$\sum F_{Total}^{M}$ represents the total muscle force acting at the knee, and $\sum F_{Agonists}^{M}$ represents the muscle force generated by the knee agonist muscle groups. The agonist muscles were the knee flexor and extensor groups for knee flexion and extension, respectively.

The point kinematics program in OpenSim tracked the attachment sites of each ligament during gait. The ligament force at each instant in time was calculated as a function of ligament length by recognising ligament stiffness, reference length, and strain. The time into the gait cycle (percent of cycle in the stance phase) for the first and second peaks of the KCF were recorded for each trial. The KCFs, knee muscle co-contraction and ligament forces during the peak KCFs were determined to facilitate the analysis of the effect of independent variables and their interactions.

### *2.4 Statistical analysis*

Dependent parameters were analysed for the chosen 96 trials with SPSS (IBM Corporation, Armonk, NY). Analysis of variance (repeated measures) was used to determine the effects (and interactions) of surface conditions, surface configurations, and uphill/downhill limb. The Greenhouse-Geisser correction was used when the assumption of sphericity was violated. Post hoc tests were performed using the Bonferroni adjustment to correct for multiple comparisons. Paired t-test was used for comparing uphill and downhill limbs. Observed power was also computed. The results were considered statistically significant when $p < 0.05$.

## 3 Results

Two peaks were observed for the KCF curves in the stance phase of gait cycle. The effect of surface condition was statistically significant for the timing of the first peak ($p = 0.001$, Table 1), but not for the timing of the second peak ($p = 0.726$). The first peak was found to occur earlier during walking on MB and WB compared with NB ($p = 0.018$ and $p = 0.010$, respectively). No statistically significant differences were found between surface configurations ($p = 0.228$ and $p = 0.912$, respectively) and limbs ($p = 0.158$ and $p = 0.909$, respectively) in the timing of the first or second peak. The second peak of the KCF was found to be larger than the first peak ($p = 0.001$). However, no independent variables or interactions had significant effects for the magnitude of the first or second peaks. The timing and magnitude of the two peaks were also similar between the uphill and downhill limbs in the lateral slant configuration.

**Table 1** The magnitude and timing of peak KCF

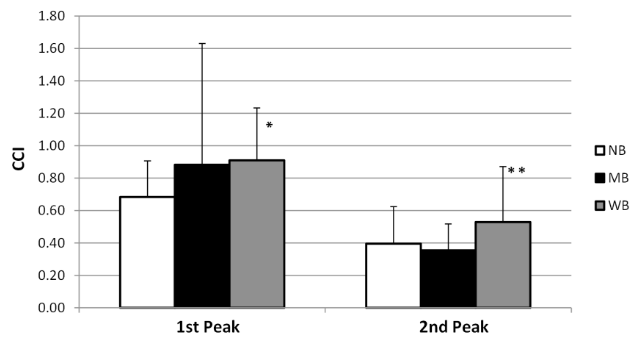
| | | *1st peak* | *1st peak time* | *2nd peak* | *2nd peak time* |
|---|---|---|---|---|---|
| Condition | No ballast | 4.57 (0.63) | 24.09 (2.88) | 6.41 (1.21) | 72.11 (3.18) |
| | Main ballast | 4.63 (0.93) | 21.63 (3.11)* | 6.08 (1.31) | 72.80 (5.31) |
| | Walking ballast | 4.51 (0.89) | 20.95 (3.76)* | 6.22 (1.14) | 72.90 (6.71) |
| Configuration | Level | 4.53 (0.82) | 21.88 (3.93) | 6.19 (1.19) | 72.67 (6.00) |
| | Slope | 4.61 (0.83) | 22.56 (3.04) | 6.29 (1.26) | 72.53 (4.37) |
| Limb | Level left leg | 4.53 (0.79) | 21.22 (4.08) | 6.14 (1.05) | 72.90 (6.19) |
| | Level right leg | 4.53 (0.86) | 22.54 (3.74) | 6.23 (1.34) | 72.44 (5.94) |
| | Slant left leg (downhill) | 4.52 (0.76) | 22.16 (3.50) | 6.33 (1.30) | 72.49 (4.37) |
| | Slant right leg (uphill) | 4.71 (0.90) | 22.97 (2.50) | 6.25 (1.24) | 72.58 (4.47) |

Notes: Peak KCF expressed in BW as mean (SD) and timing of peak KCF expressed in percentage of the stance phase in gait cycle as mean (SD). * indicates a significant difference from no ballast ($p < 0.05$).

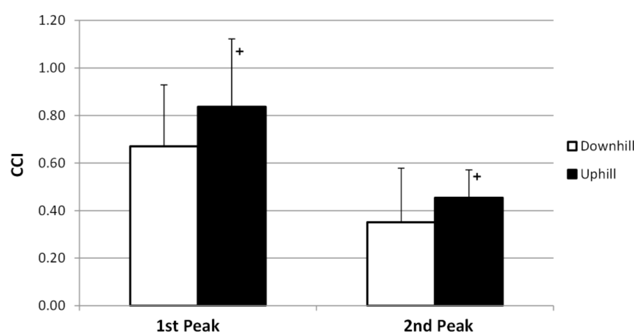
The knee muscle co-contraction was, on average, greater for the first peak of the KCF than the second peak ($p = 0.001$). The knee muscle co-contractions were significantly larger when walking on WB compared with NB in the first and second peaks [$p = 0.025$ and $p = 0.041$, respectively, Figure 3(a)] and the difference of knee muscle co-contraction also existed between two ballast conditions in the second peak ($p = 0.026$). Although it appears that the knee muscle co-contraction was higher when walking on the MB than

NB in the first peak of the KCF, the large variance in muscle co-contraction among eight subjects during walking on MB resulted in non-significant differences between these two conditions ($p = 0.792$). The knee muscle co-contraction was higher in the uphill limb than downhill limb in the lateral slant condition for both peaks [$p = 0.002$ and $p = 0.028$, respectively; Figure 3(b)].

**Figure 3** Knee muscle CCI by surface condition and limbs in peak KCFs




(a)



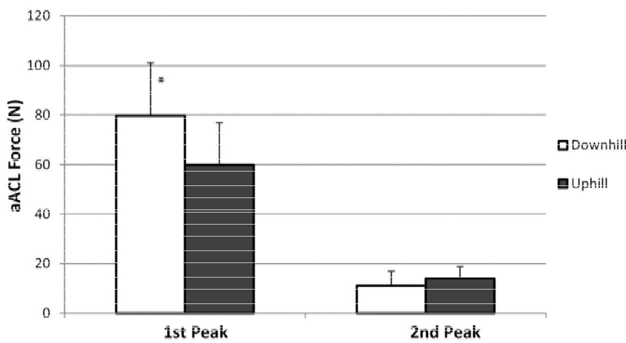

(b)

Notes: * indicates a significant difference from NB. ** indicates a significant difference from NB and MB. + indicates a significant difference between uphill and downhill limbs. Bar represent standard errors.

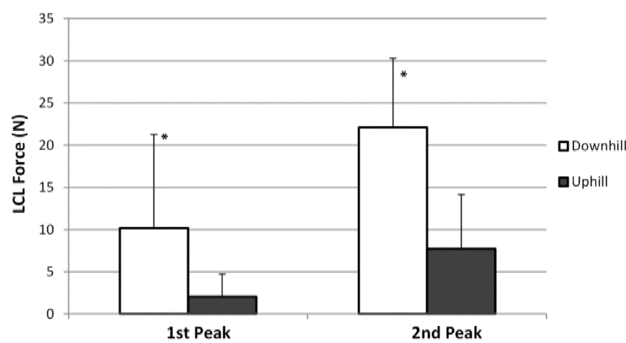
The forces generated by several ligament bundles were found to be larger for the downhill limb than uphill limb, including the aACL in the first peak [$p = 0.002$, Figure 4(a)] and LCL in both peaks [$p = 0.042$ and $p = 0.011$, respectively; Figure 4(b)]. However, the ligament forces from aMCL and iMCL were the opposite and were smaller for the downhill limb compared with uphill limb in the first and second peaks [$p = 0.017$ and $p = 0.018$, respectively; Figure 4(c) and 4(d)]. The aACL and aMCL had more force for the first peak [$p = 0.042$ and $p = 0.022$, Figure 4(e) and 4(g)] in the lateral slant configuration than level configuration, and the LCL [$p = 0.017$, Figure 4(f)] had more force for the second peak in the lateral slant configuration than level configuration. Ligament forces were significantly different between the uphill and downhill limbs: those generated by the aACL ($p = 0.001$), aMCL ($p = 0.01$) and iMCL ($p = 0.016$) were different for the first peak, and those generated by the LCL ($p = 0.006$) and aMCL

($p$ = 0.003) were different for the second peak. In other words, for the lateral slant, the ligaments responded to the surface configuration in an effort to maintain knee joint stability. These results provide support for this ligament model and its sensitivity to changes in GRF and kinematics for a lateral slant walking surface.

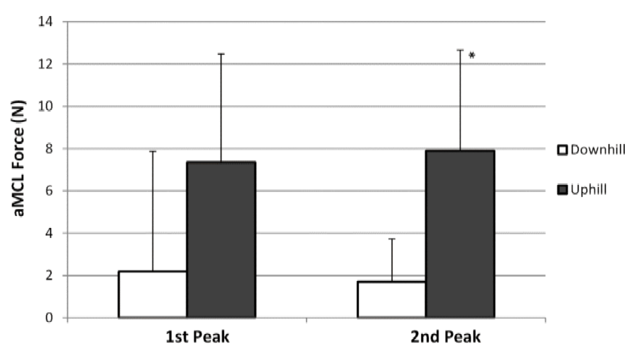
**Figure 4** Ligament forces by limbs and surface configuration in peak KCFs




(a)



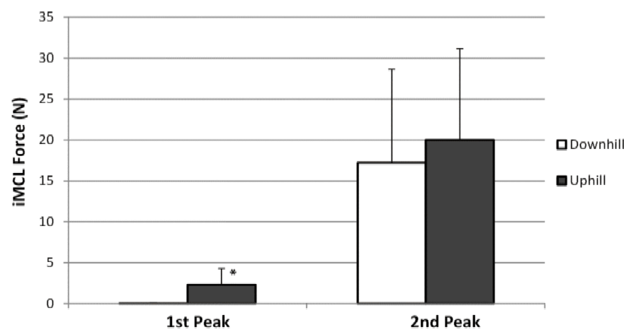

(b)



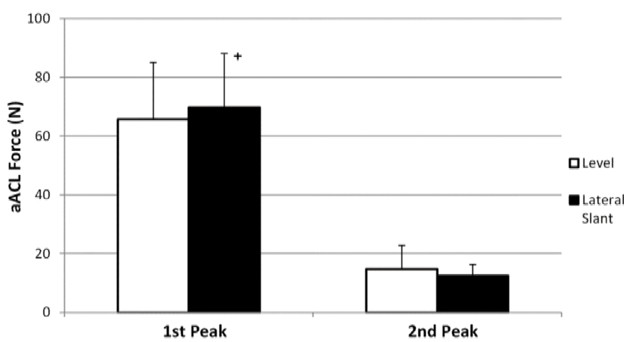

(c)

Notes: * indicates a significant difference between uphill and downhill limbs. + indicates a significant difference between level and slanted configurations. Bar represents standard errors.

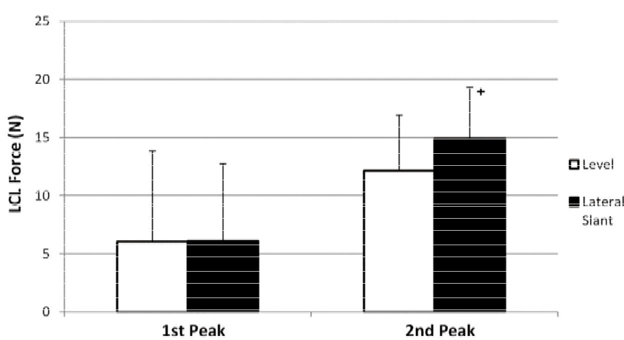
**Figure 4** Ligament forces by limbs and surface configuration in peak KCFs (continued)




(d)



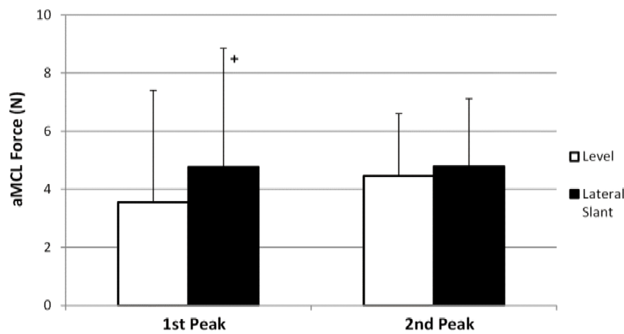

(e)




(f)

Notes: * indicates a significant difference between uphill and downhill limbs. + indicates a significant difference between level and slanted configurations. Bar represents standard errors.

**Figure 4** Ligament forces by limbs and surface configuration in peak KCFs (continued)




(g)

Notes: * indicates a significant difference between uphill and downhill limbs. + indicates a significant difference between level and slanted configurations. Bar represents standard errors.

## 4 Discussion

The first hypothesis of the study, that the peak KCF is larger during walking on ballast compared with NB, was not supported. There was no significant difference in KCF for the first or second peak when walking on MB, WB and NB. This may be explained by the decreased gait speed on MB and WB compared with NB noted in earlier research (Merryweather, 2008). A study performed by Kim et al. (2009) showed that the two peak KCFs both decreased by the reduction of walking speed. However, another study (Richards and Higginson, 2010) indicated that only the second peak, not the first peak, of KCF was influenced by walking speed. The different subject population categories for the two studies is a possible reason for the inconsistent results. Although no differences in peak KCFs were observed among surface conditions in this study, a trend was observed that the magnitude of the second peak decreased when walking on ballast compared with NB. This suggests support of the findings from Richards and Higginson (2010). No significant differences in peak KCFs between uphill and downhill limbs were found in either of the referenced studies. The similarity of the magnitude of both peak KCFs for the uphill and downhill limbs in the lateral slant configuration suggests that workers tend to use a symmetric compensatory strategy to prevent large KCF on either the uphill or downhill limbs. Even though the primary muscles involved in the two peak KCF are known, it is not clear why the timing of two peak KCFs happened earlier in stance phase during walking on ballast compared with NB (Anderson and Pandy, 2003, Sasaki and Neptune, 2010).

Other research using instrumented tibiofemoral implants indicates that peak KCF ranges from 1.8 to 3.0 body weight (BW) and other research using musculoskeletal models indicates that peak KCF ranges from 1.8 to 8.1 BW during overground and treadmill gaits (Fregly et al., 2012). In the present study, the average peak KCFs were 4.57 BW and 6.24 BW for the first and second peaks, respectively, which were higher than the reported in vivo measurements, but within the range of other musculoskeletal model predictions. The relatively high predicted peak KCF observed in the present study has several possible causes. One possible reason is that the study population was small (1 to 5 subjects) for the published studies of in vivo measurement (Taylor et al., 1998; D'Lima et al., 2007; Heinlein et al., 2009; Kutzner et al., 2010). Additionally, in previous studies with in vivo KCF, nearly all the participants were elderly subjects with osteoarthritis, which likely meant a relatively lower peak KCF due to the generally slower walking speed compared with healthy, younger adults. It is proposed that the results from the in vivo measurements relate more to elderly tibiofemoral implant patients and less to the general working population such as that used in the present study. Another possible reason for the generally higher KCF in this study is that most other musculoskeletal gait models include only muscles as force contributors and, in general, assume a 1-DOF knee joint (Sasaki and Neptune, 2010; Taylor et al., 2004). KCF may be underestimated due to the lack of ligament forces, especially ACL, in which the forces are sometimes estimated to range from 0.2 to 1.7 BW during gait (Kim et al., 2009; Shelburne et al., 2004). Additionally, the lack of recognition of knee motions in the frontal and transverse planes could cause inaccurate knee muscle forces and resulting KCF due to the possible alternative muscle excitation pattern (Kadaba et al., 1990; Xiao and Higginson, 2008).

The second hypothesis of the study, that knee muscle co-contraction are higher when walking on ballast compared with NB in the peak KCF, was supported for the WB condition for both peaks, but not supported for the MB condition for either peak. Previous research reported that muscle co-contraction was higher when subjects performed activities resulting in high KCF and that the co-contraction acted to distribute compressive forces more evenly across the articular surface (Baratta et al., 1988; Holt et al., 2003). In the context of our study, the higher knee muscle co-contraction in both peak KCFs suggests that railroad workers may have a more cautious gait during walking on WB compared with NB. Why this is not also the case for MB is unclear. This finding was in agreement with previous muscle co-contraction results measured by EMG signals (Wade et al., 2010). Knee muscle co-contraction was also found to be significantly higher for the uphill limb than downhill limb for the first and second peaks of the KCF in this study. Existing literature suggests that higher muscle co-contraction may result in increased risk of acute or chronic joint injury and muscle fatigue, an intrinsic factor contributing to slips and falls (Parijat and Lockhart, 2008; Bentley and Haslam, 2001). It is therefore proposed that the uphill limb may have more risk for knee injury and development of knee osteoarthritis than the downhill limb.

The third hypothesis of the study, that forces in the medial and lateral collateral ligaments are higher for uphill than for downhill limbs during peak KCF when walking on the lateral slant surface, was supported for the MCL but not the LCL. The MCL (iMCL and aMCL) generated more force for the uphill than the downhill limb and the LCL generated more force for the downhill than the uphill limb in both peaks. This is expected since resisting knee valgus and varus stress are the main function for MCL and LCL, respectively. These findings agree with the observation that knee adduction angles were larger for the downhill limb compared with the uphill limb in the first and second peaks of the KCF. It was also found that the forces generated by aACL were larger in the first peak and the LCL forces were larger in the second peak for the lateral slant [Figures 4(e) and 4(f)]. Previous knee ligament function analysis in the current gait model indicated that ligament force increased for aACL, but decreased for LCL with increased knee flexion (Xu, 2013). It should be noted that there was also an interaction effect of configuration by limb for some ligaments (ACL, MCL and LCL). This is not surprising since the right and left limbs were always the uphill and downhill limbs in the lateral slant configuration, respectively.

We acknowledge some limitations in this study, most notably the relatively small sample size, which in most cases resulted in a statistical power below 0.5. This might also limit the identification of some of the main effects of the independent variables, which reduces the ability to confidently generalise the study results to the entire population of railroad workers. Second, the knee proximodistal and anteroposterior translation were defined as a function of knee flexion in the present model. However, the effect of this knee translation-flexion function on ligament length is unclear. A sensitivity analysis for the ligament in the present model suggests that a 10% change in the ACL length could cause ligament forces to vary by as much as 200 N when strained above 3%.

In summary, this study examined joint loading, muscle co-contraction activity and ligament forces at the knee in the two peaks of the KCF while walking on level and slanted ballast. Results indicate that at the times of the two peak KCFs

1. the effects of limb, surface condition and configuration were not significant for two peak KCFs, but the second peak KCF tended to be lower for both ballast conditions compared with NB
2. the timing of the first peak KCF occurred earlier in the stance phase during walking on ballast than NB
3. the knee muscle co-contraction was higher when walking on WB than NB
4. the forces generated by knee collateral ligaments were significantly different between the uphill and downhill limbs when walking on the lateral slant condition.

It appears that the walking gait of railroad workers may include compensatory mechanisms, such as reducing walking speed, increasing muscle co-contraction, and optimising ligament forces, to prevent high peak KCF, stabilise the knee joint, and reduce the possibility of knee injuries and/or falls during walking on ballast.

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