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# Reaction Moment at the L5/S1 Joint During Simulated Forward Slipping With a Handheld Load

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*The purpose of the study was to investigate the effects of load on the net moment response at the L5/S1 joint during simulated slip events. Six young individuals were instructed to take one step with a handheld load. Sudden floor movement was randomly introduced to simulate unexpected slips. Different loads conditions (0%, 10%, 20%, 30% of body weight) were introduced at random. Three-dimensional net moments at the L5/S1 joint were computed via downward inverse dynamic model. Peak joint moment generated at 30% load level was found to be significantly higher compared to no-load condition. No peak moment differences were found among no-load, 10% or 20% load levels. Additionally, the findings from this study indicated a flexion-dominant net L5/S1 joint moment pattern during motion phase associated with slip-induced falls.*

back injury load carrying slips and falls reactive recovery

## 1. INTRODUCTION

In 2011, the U.S. Bureau of Labor Statistics reported that back injury was the most frequent cause for nonfatal injuries and illnesses involving days away from work in the USA [1]. Overexertion and bodily reaction were the most frequent cause leading to injury or illness in 2011 [2]. Material movers/transportation occupations had the highest frequency of overexertion and they were employed in the wholesale/retail trades [3]. Low back injuries were often associated with over-exertion during manual materials handling (MMH) tasks [4].

According to Courtney and Webster, one workers' compensation provider claimed that cost ratio[1] for ruptured disc due to the fall to the same level was highest (13.3) among many injury claims [5]. The Bureau of Labor Statistics in 2011 indicated that among 225 980 back injuries, 35 580 injuries were related to slips, trips, and falls. In addition, out of 415 800 overexertion and bodily reaction injuries, 166 720 cases affected the back; bodily reaction included slips and trips without falls, but also other injuries due to bodily motion or reaction[2]. Mattila, Kaustell, Rautiainen, et al. indicated that 45% of nonfatal injuries were due to slips,

[1] Cost ratio is the ratio of the average cost of the particular injury to the average cost of all injuries for that particular class of falls.

[2] Data from http://stats.bls.gov.

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trips, and falls in an open-field vegetable production environment [6]. Balance perturbation or falls caused by a slippery or unstable floor surface while carrying a load can make this problem even worse [7]; occupational load carrying is one of the most frequent tasks that MMH workers perform at their job site, e.g., when delivering mail and stocking.

Studies on human responses to unexpected gait perturbations mainly focus on corrective strategies of lower extremities. Hip muscles are the most important muscles associated with maintaining balance while experiencing simulated slip events [8]. The ankle joint, on the other hand, acts as a passive joint during fall or reactive-recovery trials [9]. Meanwhile, the corrective strategy includes increased knee flexion moment and hip extensor moment [9]. However, the added weights due to the nature of MMH work may complicate the reaction mechanisms, and consequently the injury characteristics in the back. Epidemiological studies have indicated that sudden loading to the trunk is associated with acute low back pain and may be a primary risk factor for chronic low back pain development [10, 11]. Floor surface inclination will also contribute to low back loading while carrying or lifting a load [12]. Unexpected gait perturbations can be dangerous to the lumbar spine because of the rapid corrective movements needed to regain balance. For example, trunk acceleration increased significantly during unexpected perturbation, such as slipping compared to that during normal gait [13]. In addition, the most common site of injury caused by slipping was found to be the lumbar spine, more frequent than ankle and knee injury [10].

Gait control of walking under different load-carrying conditions has been explored by numerous researchers [14, 15, 16, 17]. In a simplistic walking model, adding a load can alter the inertial properties of the model, such as shifting the body center of mass position and adding overall weight. When adding the load in the back, e.g., in the case of firefighters carrying the air bottle, increased load weight has been found to result in a decrease in gait performance and an increase in risk of tripping [14]. Although information about the factors influencing the likelihood of slips while carrying a load is available, information about the back reaction mechanisms associated with a slip event is lacking. An understanding of the mechanism will provide an insight into back injury mechanisms while slipping or falling.

The objective of the current study was to investigate the effects of load carrying on three-dimensional (3D) L5/S1 joint moment response during unexpected step perturbation, such as simulated forward slipping. Although the current study was not evaluated under a realistic environment, understanding the responses to the unexpected perturbations should help scientists to understand possible injury mechanisms during real slip events. The present study hypothesized that higher load level would result in higher demand on body corrective activity (i.e., higher L5/S1 joint moment in the current study), and thus increase the risk of low back injury occurrence.

## 2. METHODS

### 2.1. Subjects

Six young healthy adults, whose mean (*SD*) age, weight, and height were 27 (1.29) years, 67.7 (3.81) kg, and 167.88 (1.41) cm, respectively, participated in the study. Informed consent was approved by the Institute Review Board of Virginia Tech and was obtained from all participants before any data collection.

### 2.2. Procedure and Equipment

The participants were instructed to walk one step forward onto the force-plate (Bertec #K80102, type 45550-08, Bertec, USA), which was designed to produce a sudden forward slide (a total distance of 25 cm at 60 cm/s). All participants were provided with identical shoes, but different size. Weights were added and fixed in a box that was shielded with 10 cm of polystyrene foam. The cushioned box was handheld in front of the participants. All four load levels (0%, 10%, 20% and 30% of respective body weight) were presented to each participant in random order. For 0% levels, the participant's arms were at the sides without holding the load. During each trial, from the standing posture with the weight in front, participants were asked to make a natural forward step onto a platform, which would either produce a sudden forward slide or produce no slide. The simulated slip events (i.e., sliding motion) were introduced randomly. Immediately after the heel contacted the platform, when the vertical ground reaction forces exceeded 7 N, the platform began to travel away from body and stopped immediately after traveling 25 cm at 60 cm/s. Twenty spherical reflective markers were attached onto the bony landmarks of the participant's upper body and trunk to perform downward inverse dynamics. The detailed marker configuration was as follows: left/right distal head of the third metacarpal, left/right radial styloid, left/right ulnar styloid, left/right medial–lateral humeral epicondyles, left/right acromioclaviculare, left/right ear, top of the head, left/right acromion, L5/S1 joint, left/right heel. In addition, three reflective markers were attached on top of the load to record load position. A six-camera infrared system (ProReflex MCU 120, Qualisys Medical, Sweden) was used to capture marker position data at 120 Hz and the data were low-pass filtered by a zero-phase fourth-order Butterworth filter with a cut-off frequency of 6 Hz. A fall-arresting harness was employed to prevent the participant's body from striking the ground [18]. Athletic shoes of one type were provided to each participant.

### 2.3. Measurement

Downward 3D inverse dynamics was applied to obtain L5/S1 net joint moments. Computational details for 3D inverse dynamics can be found in a previous publication [19].

Briefly, initial input information for downward inverse dynamics included kinematic measurements of the upper extremity and trunk segment, external forces (e.g., handheld load) and point of applications (e.g., hand center of mass) at the distal end (e.g., hands), and an estimate of body segment parameters. The local co-ordinate system was constructed through the Gram–Schmidt orthogonalization process [19, 20]. An expanded equation of motion was applied to derive 3D joint moments, starting from the wrist joints, followed by elbow, shoulder and L5/S1 joints. The entire trunk was modeled as a single rigid segment connected to the pelvic segment through the L5/S1 joint.

L5/S1 net joint moments were expressed based on trunk reference frame [19]. Joint moments were normalized as a percentage of participants' weight.

L5/S1 joint moment responses and anterior–posterior velocity and acceleration of the sliding platform were analyzed during motion phase (from motion-start until motion-end). Platform motion-start was defined as the first instant that forward platform velocity occurred after heel contact. Platform motion-end was defined as the time when the forward platform velocity became zero.

Ensemble averages of flexion–extension moment at the L5/S1 joint were illustrated at each load level. Motion phase was extracted and scaled to 100% of relative time. One-way within-subject analysis of variance (ANOVA) was performed to test the effects of load level on peak joint moments in three reference planes (sagittal, transverse, and frontal). Post hoc (Student Newman–Keuls) test was performed to further investigate differences among the load levels (0%, 10%, 20%, and 30%). A significant level of $p \leq .05$ was used throughout the analyses. Descriptive and inferential statistical analyses were performed with JMP version 7 [1].

## 3. RESULTS

### 3.1. Net L5/S1 Joint Moment in Sagittal Plane

Figure 1 illustrates the ensemble averages of net L5/S1 joint moment in the sagittal plane of four load levels. Generally, sagittal moment responses at the L5/S1 joint were flexion dominant across all three load levels. Maximum joint moment typically

[1] http://www.jmp.com/

occurred ~30%–40% of relative time after motion-start. After motion-start, flexion moment decreased until 40% of motion phase and then increased again. Similar patterns were observed for all the participants at all four load levels.

There was a significant load level effect ($p$ = .005) on maximum sagittal joint moment response. The fitted model was deemed acceptable, as the adjusted $R^2$ was found to be .83 for lateral bending, .80 for internal–external rotation, and .75 for flexion–extension. In addition, the residuals of the fitted model appeared to behave randomly across load levels in all three reference planes; the lag plots of the residuals in all levels suggested that the residuals were independent. This suggested that the model fitted the data well. The post hoc test indicated that the only significant difference was between 30% load level and the other three load levels (Table 1). Therefore, significantly more maximum flexion moment was generated at the L5/S1 joint while carrying a 30% load than carrying no load, 10%, or 20% load, after the participant's balance was being perturbed by sudden platform sliding motion.

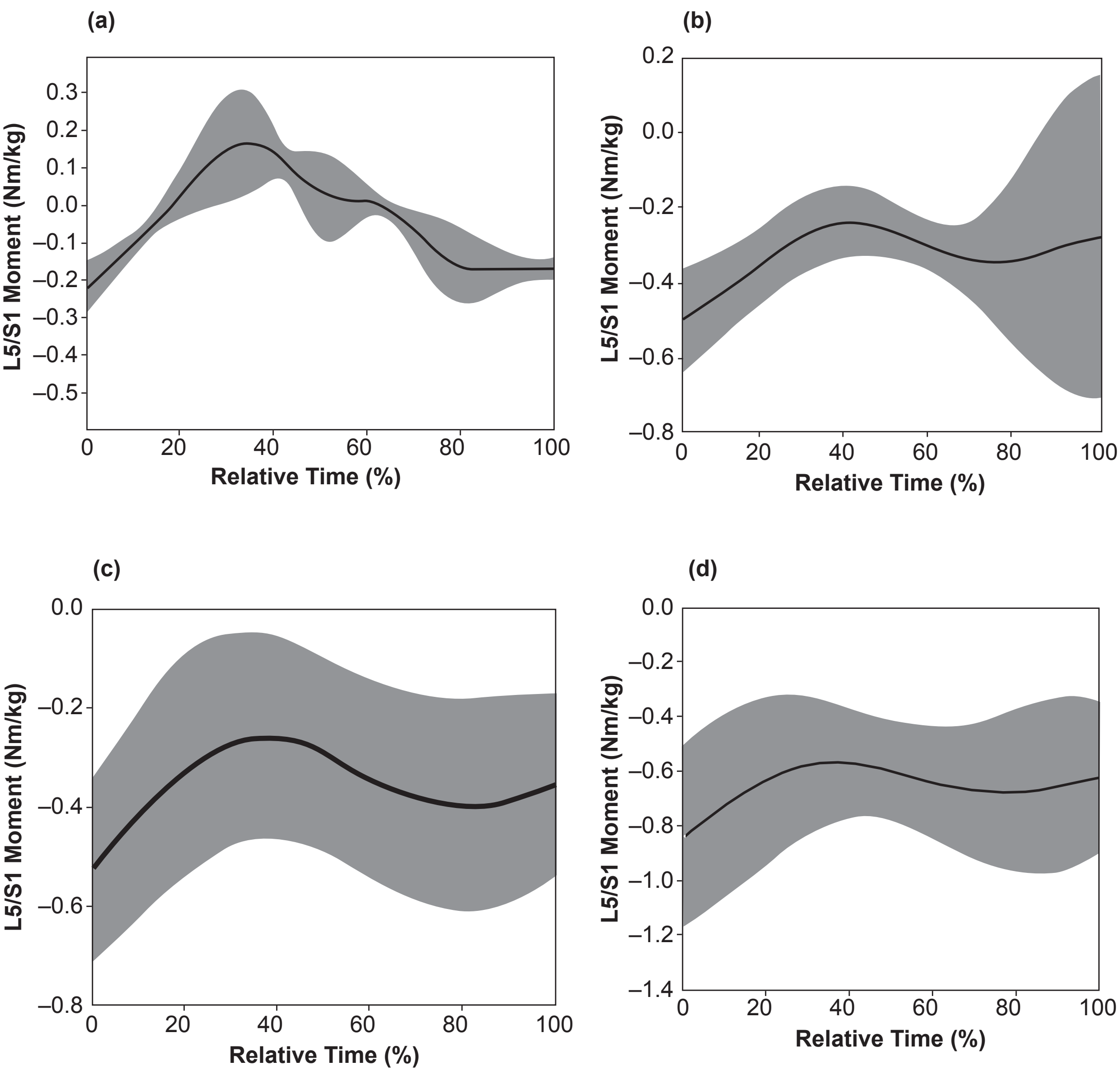


**Figure 1. Ensemble average of flexion–extension net L5/S1 moment at (a) 0%, (b) 10%, (c) 20%, and (d) 30% load level.** *Notes.* + = extension; – = flexion; 0%, 100% of relative time ($x$ axis) = plate motion start and end, respectively. Solid curve represents ensemble average joint moment, shaded area indicates ± 1 *SD*.

**TABLE 1. Peak Moment at L5/S1 Joint, *M (SD)***

| Load Level (%) | Lateral Bending | Int–Ext Rotation | Flexion–Extention | *p* |
|---|---|---|---|---|
| 0 | 0.13 (0.02) | 0.12 (0.08) | 0.26 (0.11) | .006 |
| 10 | 0.29 (0.14) | 0.28 (0.19) | 0.52 (0.13) | .035 |
| 20 | 0.29 (0.12) | 0.24 (0.09) | 0.54 (0.18) | .027 |
| 30 | 0.50 (0.16) | 0.60 (0.76) | 0.86 (0.31) | N/A |

*Notes.* Int–Ext = internal–external; *p* indicates pairwise comparison results between 30% load and the other 3 load levels; N/A = not applicable.

### 3.2. Net L5/S1 Joint Moment in Frontal and Transverse Plane

Compared with joint moment generated in the L5/S1 sagittal plane, net frontal and transverse L5/S1 moment did not show a clear recognizable pattern. Peak joint moment magnitude was also generally smaller than the results obtained from sagittal plane.

There was a significant load level effect ($p$ = .001) on frontal net moment at the L5/S1 joint. The post hoc test indicated that peak joint moment was significantly greater at load level of 30% (0.50 ± 0.16) than at the no-load condition (0.13 ± 0.02) (Table 1).

No significant load level effect ($p$ = .511) on transverse L5/S1 joint moment was evident. Therefore, a different level of load carrying (including the no-load condition) was not found to affect L5/S1 transverse loading significantly.

## 4. DISCUSSION

The objective of this study was to quantify the effects of load on 3D net moment responses at the L5/S1 joint during simulated slip events.

The results from this study suggest that light loads (e.g., 10% and 20%) in front of the trunk may not have a significant effect on net joint moments at the L5/S1 joint during an unexpected slip, although high loads significantly increase the low back loading by over 300% compared to the no-load condition. These results suggest that the 10% and 20% load level can be recommended for a light-weight load, while 30% can be a heavy-weight load, considering the fact that Cathcart, Richardson and Campbell recommend 40% of body weight as the maximum carrying weight [21]. Between 20% and 30% of body weight can be a borderline weight that can be carried in front without a significant effect on net joint moment at the L5/S1 joint.

The current results indicate that load level effect is only evident at 30% of body weight level where maximum L5/S1 moment is 187.5% higher in sagittal plane and 284% higher in frontal plane compared to the no-load condition. Such high flexion peak joint moment at 30% load level is comparable to peak moment during 40° forward-bending lifting tasks reported by Larivíere and Gagnon [22]. Although only heavy load carrying would complicate back injury mechanism, previous studies have found that a light load influenced some gait characteristics of adults. Kim and Lockhart suggested that carrying 10% of body weight appeared to influence heel contact velocity and step length, but not the slip propensity (e.g., friction demand at shoe–floor interface) [23]. However, Myung and Smith found only load level at 40% of body weight to have significant effect on the stride length [16]. They suggested that the load effect found only at high load level could be plausibly explained by the adaptive gait control mechanism which tolerated lighter added weight but not heavier added weight. These results from previous studies and the present study further suggest that carrying a light-load will possibly not complicate gait mechanisms in association with slips or back injury mechanisms caused by a slip.

In the present study, carrying a load had no effect on peak L5/S1 joint moment in the transverse plane. There were no statistically significant peak moment differences between different load levels and no-load conditions. Table 1 shows that peak transversal moment at 30% load level had a large standard deviation, which was higher than its mean value. Such high variation of L5/S1 moment response in the transverse plane may have contributed to a nonsignificant load effect finding. Similarly, large standard deviations found in both lateral bending and flexion–extension could have contributed to nonsignificant effects between different load levels, e.g., 10%, 20% and 30%. Overall, except for sagittal moment response, there was no clear pattern among all the participants for frontal and transverse moment generation, during unexpected slip events. These results suggested the existence of individual-specific moment generation strategies for balance adjustment in the frontal and transverse planes.

Overall, flexion moment decreased until ~35%–40% of motion phase after the initial high peak moment and increased thereafter. Similar patterns were observed for all participants at all three load levels. Thirty to 40% of motion phase corresponds to 0.15–0.17 s in time after the motion-start; 25 cm sliding distance at 60 cm/s (25 cm/[60 cm/s] = 0.42 s, 0.42 × 0.35 = 0.15 s and 0.42 × 0.40 = 0.17 s). Total slip distance at 0.35 and 0.42 s corresponds to ~8.72–10.00 cm, respectively (0.35 × 25 cm or 0.40 × 25 cm). The participants seemed to react rapidly to the sudden simulated slip and react again at ~8.50 or 10.00 cm slip distance. This common pattern can lead to finding the slip distance that can be considered as a dangerous slip. In the present study, the results suggested that the participants sensed the threat from slipping at a slip distance of ~8.50–10.00 cm after the initial threat (i.e., motion-start). This indicates that slip distance greater than ~8.50 cm is a threshold for detecting a dangerous slip. Individuals' ability to regain balance at this slip distance can possibly dictate whether a person falls or recovers from a dangerous slip. Although further studies are necessary to test this hypothesis about the threshold slip distance, the possible findings may already be seen in previous studies. Liu and Lockhart evaluated reactive-recovery mechanisms at lower extremities and suggested that reactive-recovery often started at 25% of the stance phase [24]. This finding may suggest that an individual does not perceive the initial slip as a dangerous slip, and thus does not see a need to recover from the slip. But, an individual senses a danger or a need to respond to a slip distance at ~25% of the stance phase. The stance phase lasts ~0.50 s, not under 0.40 s [25] and average sliding heel velocity is 65–85 cm/s [27, 28]. The sliding velocity of 65 and 85 cm/s corresponded to 8.13 and 10.63 cm, respectively, in the stance time of 0.50 s in the current study. The summarized data from previous studies [24, 25, 26] agreed with the results from the present study. This further suggests that recovering from a slip would depend upon the capability to develop appropriate reactive motions at the slip distance of 8.00–10.00 cm, not upon the definite length of the slip distance as Grönqvist suggested [28].

Certain limitations existed in the current study. The relatively small sample size limited the statistical power to detect other relevant findings although the study detected significant differences between 0% and 30% loads. In addition, simulated sudden floor motion could not be fully compatible with the situation one encounters during a slip induced by a slippery surface although this type of floor motion was reasonable to simulate unexpected slips caused by unstable floor constructions.

In conclusion, only heavy load carrying (30% of body weight) was found to increase peak L5/S1 moment generation, and thus increased the risk of suffering low back injury during an unexpected slip event. Further studies should focus on validating general gait characteristics under unexpected slips induced by both sudden floor motion (used in the current study) and a slippery surface, and of the capability of regaining balance at the threshold slip distances (i.e., 8.00–10.00 cm).

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