Recovering from Laboratory-Induced Slips and Trips Causes High Levels of Lumbar Muscle Activity and Spine Loading

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Abstract

Slips, trips, and falls are some of the most substantial and prevalent causes of occupational injuries and fatalities, and these events may contribute to low-back problems. We quantified lumbar kinematics (i.e., lumbar angles relative to pelvis) and kinetics during unexpected slip and trip perturbations, and during normal walking, among 12 participants (6F, 6M). Individual anthropometry, lumbar muscle geometry, and lumbar angles, along with electromyography from 14 lumbar muscles were used as input to a 3D, dynamic, EMG-based model of the lumbar spine. Results indicated that, in comparison with values during normal walking, lumbar range of motion, lumbosacral (L5/S1) loads, and lumbar muscle activations were all significantly higher during the slip and trip events. Maximum L5/S1 compression forces exceeded 2700 N during slip and trip events, compared with ~1100 N during normal walking. Mean values of L5/S1 anteroposterior (930 N), and lateral (800 N) shear forces were also substantially larger than the shear force during the normal walking (230 N). These observed levels of L5/S1 reaction forces, along with high levels of bilateral lumbar muscle activities, suggest the potential for overexertion injuries and tissue damage during unexpected slip and trip events, which could contribute to low back injuries. Outcomes of this study may facilitate the identification and control of specific mechanisms involved with low back disorders consequent to slips or trips.

Keywords: gait perturbation; spine loading; kinematics; kinetics; muscle activity

1. Introduction

Falls on the same level are among the leading causes of accidental injuries and fatalities across many workplaces. Such falls, many subsequent to slips and trips, were ranked as the second most frequent cause of disabling workplace injuries after overexertion—accounting for 19.2%

of these injuries and imposing direct costs of $11.2 billion in the USA (Liberty Mutual, 2018). The scope of this problem is substantial, with 300,000 slip-and-fall injuries involving days away from work occurring annually in the USA (Bureau of Labor Statistics, 2017). Even if a fall is averted, slips and trips without falling (STWF) account for 3.9% of back injuries (Liberty Mutual, 2018).

In response to slip or trip events, an adequate recovery effort is necessary to regain balance. In some cases, the required effort can be substantial, involving rapid and large body movements that could lead to large loads on certain body tissues and subsequent injury. Most existing biomechanical analyses of slip and trip events have focused primarily on the lower extremity, demonstrating large knee flexion and hip extension moments when recovering from a slip (Cham & Redfern, 2001; Yang & Pai, 2010) and large ankle and hip extension moments in recovering from a trip (Pijnappels et al., 2005).

Sudden loading to the trunk also occurs during STWF and has been associated with low-back pain (Manning et al., 1984; Manning & Shannon, 1981b; Rohrlich et al., 2014), yet there is only limited evidence on lumbar spine kinematics and kinetics occurring during/after unexpected gait perturbations. Large horizontal accelerations of the trunk occur during efforts to regain balance after slipping and tripping (Hirvonen et al., 1994). Additionally, increased L5/S1 moments were found when recovering from slip events simulated via platform displacements during load carriage (Liu et al., 2014). Rashedi et al. (2012) reported pilot results ($n = 6$) on lumbar responses to an unexpected slip, which suggested higher lumbar muscle forces, along with higher L5/S1 reaction moments and internal forces (i.e., compression and shear forces), compared to those found during unperturbed gait.

To better understand the potential contributions of STWF effects to low-back pain, the purpose of the current study was to quantify low back muscular reactions, along with lumbosacral kinematics and kinetics, during recoveries from both unexpected slips and trips. Based on our prior results, we hypothesized that both types of events would cause large changes to lumbar angles and muscle forces, along with increases in L5/S1 kinetics.

2. Materials and Methods

2.1. Participants

A convenience sample of 12 participants (6M, 6F) was recruited from the local university and nearby community. Respective means (SD) of age, stature, and body mass were 24 (2.4) years, 176 (5.8) cm, and 70 (6.9) kg for males; and 24 (2.8) years, 166 (4.5) cm, and 57 (5.2) kg for females. All participants completed informed consent procedures approved by the Virginia Tech Institutional Review Board prior to participation in the study. Participants self-reported no current or recent (past 12 months) history of musculoskeletal disorders/injuries or low-back pain and being physically active (i.e., exercising at least two times per week). All self-reported being right-leg dominant.

2.2. Experimental Design and Procedures

A repeated-measures design was implemented, requiring participants to complete several walking trials, along with trials involving an unexpected slip or trip. Participants repeatedly (~5 trials) walked back and forth, at a self-selected, comfortable speed, on a 15.5 ×1.5 m track (Figure 1). This track was covered with commercial tile and had two embedded force platforms (AMTI OR6-7-2000, Watertown, MA, US). During walking trials, the starting point was adjusted to ensure participants stepped appropriately in the perturbation-inducing area and over the platform. Two perturbation events were introduced at a random time: an unexpected slip and an unexpected trip (Figure 1). The order of trip and slip perturbations was counterbalanced across the participants, and these perturbations were induced as in earlier work (Allin et al., 2020; Allin et al., 2018; Garman et al., 2016). Slips were induced by covering the surface with vegetable oil. Trips were induced by a manually operated “trap-door” mechanism; when released, a spring rapidly rotated a 5 cm obstacle in front of the foot during the swing phase of gait. To ensure participants’ safety, each wore a fall-arrest harness that was connected to an overhead rail. All participants wore a consistent type of athletic shoe. To distract participants from the exact location and time of the trip and slip events, they were asked to count colored circles displayed on a screen at the end of the walking track during all trials (Parijat & Lockhart, 2008) and to arrange some colored papers at the end of the track. All perturbations (slips/trips) were applied to the dominant foot, which was the right foot for all participants. Most participants experiences only a single trip and slip. In a few cases (two slips and one trip), trials needed to be repeated after a random number of unperturbed trials, because participant foot placement was not on the force platform.

Figure 1 about here

2.3. Instrumentation and Data Processing

Muscle activity was measured via surface electromyography (EMG), using pairs of pre-gelled, bipolar, Ag/AgCl electrodes (AccuSensor, Lynn Medical, MI) with a 2.5 cm inter-electrode spacing. Electrodes were placed bilaterally in the lower lumbar region, including three trunk flexors – internal oblique (IO), external oblique (EO), rectus abdominis (RA) – and four trunk extensors – iliocostalis lumborum pars lumborum (ILL), multifidus (MF), longissimus thoracis pars lumborum (LTL), and longissimus thoracis pars thoracis (LTT). Electrode placement followed procedures described earlier (Jia et al., 2011). Raw EMG signals were sampled at 1 kHz using a telemetered system (TeleMyo Desktop DTS, Noraxon, AZ, USA), band-pass filtered (20-500 Hz), rectified, and low-pass filtered at 5 Hz (2nd order Butterworth, bidirectional) as in prior work (Jia et al., 2011; Lemos et al., 2015).

Resting values of muscle activation were recorded in prone and supine positions. Maximal levels of voluntary muscle activation were measured from three replications, each of maximum voluntary isometric contractions (MVICs) in trunk flexion/extension, clockwise and counterclockwise axial rotation, and left/right lateral bending. Participants were asked to reach their maximum level of muscle activity with rapid exertions. In contrast to more traditional ramped efforts, these procedures were used to better mimic the dynamic movements involved in recovering from an induced slip or trip. During MVICs, participants stood in a customized fixture attached over a force platform (AMTI OR6-7-2000, Watertown, MA, USA), similar to the approach described by Granata et al. (1996). In the MVICs, participants pushed against a pad attached to the wall that made contact at the upper torso. The pelvis and lower extremities were partially immobilized movements using straps at the knees and pelvis. For axial rotation MVICs, shoulder straps were added to enable the participants to exert torsional moments (Jia et al. 2011) All exertions were performed in an upright posture except for the extension trials, which were conducted while the trunk was flexed ~20°. Normalized EMG (nEMG) values were obtained from the walking and perturbation trials, using individual maximal and minimal values.

Segmental kinematics were obtained from 26 retroreflective markers attached over bony landmarks at the pelvis (bilateral anterior/posterior sacral iliac spine) and trunk (C7 and T10 spinous process, incisura jugularis, xiphoid process, and bilateral acromial process). Marker locations were collected at 100 Hz using a 7-camera motion-capture system (Vero, Vicon, Denver, CO, USA), and subsequently low-pass filtered (9 Hz cut-off; 2nd order Butterworth; bidirectional). Following the definition of the pelvic and trunk coordinate systems recommended by Davis et al. (1991), lumbar angles (relative to the pelvis) were derived using the y-x-z rotation sequence. Specifically, the global x axis was the walking direction (lateral bending axis), the y axis was to the right (flexion/extension axis), and the z axis was upward (axial rotation axis).

2.4. Biomechanical Modeling

An anatomical model developed in the AnyBody™ musculoskeletal modeling system (v5.1, AnyBody Technology, Aalborg, Denmark) was used to estimate lumbar muscle geometry and kinematics. Initial insertions, via points, and the origins of a total of 76 muscle fascicles were adopted from pre-defined values in the AnyBody repository and scaled based on each participant's anthropometry. The AnyBody model was driven using marker data and provided outputs that included lumbar kinematics (i.e., lumbar angles relative to pelvis) and the lengths, velocities, and moment arms of the muscle fascicles. These outputs were employed as input to a dynamic, 3D, EMG-based model of the lumbar spine, along with nEMG and participant anthropometry. This model has been described in detail elsewhere (Jia et al., 2011) and been implemented for a range of task demands (Kim et al., 2011, 2012; Kim et al., 2018). Using this model, lumbar muscle forces were calculated, along with L5/S1 reaction moments and internal forces (i.e., compression and shear forces).

2.5. Statistical Analysis

One-way, repeated measure analyses of variance (ANOVAs) were conducted to assess the effect of walking condition (normal walking, slip, and trip) on the dependent measures, with gender included as a blocking effect. These measures included the ranges of lumbar angles, peak nEMGs, and peak L5/S1 moments and forces. Post-hoc paired comparisons were performed using Tukey's Honest Significant Difference (HSD) where relevant. Transformations were applied (e.g., log and sqrt) to meet ANOVA assumptions; for ease of interpretation, however, summary results are presented in the original units. Partial eta-squared (η_p^2) was used to assess effect sizes (Cohen, 1988). Statistical analyses were performed with JMP Pro 14.0 (SAS Institute Inc., Cary, NC, USA), using the REML method, and with statistical significance set at $p < 0.05$. Representative kinematic, normalized lumbar muscle activity (nEMG), and kinetic information for one participant are presented below (Figure 2) for one slip event and one trip event involving the right leg. For illustrating responses, the time of heel contact on the force platform was identified (i.e., when the vertical recorded force exceeded 25 N).

3. Results

A summary of the ANOVA results and descriptive statistics are provided in the Appendix for both kinematic and kinetics (Tables A1 and A2), and for muscle activity (Tables A3 and A4). Notably, walking condition had a significant effect (all p-values < 0.0009) on all dependent measures, and there were no significant main effects of gender. Subsequent sections present further details about the different categories of dependent measures.

3.1. Lumbar Angles

The ranges of axial rotation, flexion/extension, and lateral bending motion of the lumbar spine (relative rotations of the torso vs. pelvis) were larger after the slip or trip perturbation, compared to normal walking (Figure 2). After both types of perturbation (~0.5 sec following right heel contact), there was a large flexion motion for the individual illustrated. Lateral bending to the left and counterclockwise rotation was also evident. Such triaxial rotations, which were larger than during normal walking, were observed consistently across participants following perturbations (Figure 3, Table A2). Effects of slips and trips on lumbar angles were generally similar, though slip events caused significantly (p = 0.0103) larger extension (mean = ~33°) than trip events (mean = ~25°). There was a significant interaction effect of walking condition × gender on lateral bending angles, evident as a larger effect of a slip vs. trip among females but with more consistent effects among males. Despite general similarities during perturbation recoveries, we also qualitatively observed substantial individual differences in the detailed kinematic responses (e.g., delays until peak kinematic and kinetic values).

Figure 2 about here

Figure 3 about here

3.2. Lumbar Muscle nEMG

Muscle activity levels were considerably lower during normal walking than following a slip or trip (Figure 4, Table A3). All muscles exhibited a substantial increase in activation soon after heel strike, reaching maximum levels after ~ 0.4 sec. Post-hoc analyses indicated that muscle activity was significantly larger bilaterally compared to the normal walking condition. Many muscles approached respective maximum activation levels (> 80% MVIC) during the post-perturbation recovery from a slip or trip (Table A4). **Means of the maximum normalized muscle activations were all significantly different between normal walking and the perturbation conditions (i.e., slips and trips), yet did not differ between the two perturbation types (Figure 5).**

Figure 4 about here

Figure 5 about here

3.3. L5/S1 Moments and Forces

Peak values of L5/S1 moment components were significantly larger during the perturbation trials than in normal walking (Figure 3, Tables A1 and A2), with a representative example shown in Figure 6 (top row). During slip and trip trials, the maximum flexion moment increased by roughly 5 and 4 times, respectively. Peak lateral bending moments were roughly twice as large during perturbations, and nearly 3- and 4-fold increases in peak axial moments were observed during the slip and trip trials, respectively. There were no significant differences in peak moments between slip and trip trials.

Figure 6 about here

For the representative participant (Figure 6, bottom row), both a slip and trip caused compression and posterior shear forces that peaked ~0.5 s after heel strike. The absolute values of peak forces were considerably higher than the corresponding values during normal walking. In contrast, lateral shear forces were comparable across conditions for this participant. Across participants, there were significant differences in all three spine forces between conditions (Figure 3, Table A1). Compressive forces were ~140% higher during both slips and trips than during normal walking, and shear forces during the perturbations were respectively ~270% and 160% larger in the anterior/posterior and lateral directions (Table A2). There were no significant differences in peak forces between slip and trip trials.

4. Discussion

Findings of this study were consistent with our hypothesis: in comparison to normal walking, gait perturbations increased the range of lumbar motions (Tables A1 & A2) and muscle activations (Tables A3 & A4) and caused higher L5/S1 reaction moments and forces (Tables A1 & A2). However, some consideration is warranted regarding spine forces, given that they were derived from a model. Our EMG-based model used measured kinematics and incorporated actual muscle activation levels to predict lumbar muscle forces and L5/S1 reaction forces. Although this model has been evaluated for other tasks, including diverse manual handling activities (Jia et al., 2011), further investigation is necessary to validate the findings of the current study, especially since slip and trip events have a more dynamic nature compared to the conditions in our earlier investigations.

Nonetheless, outcomes of the current work during normal walking are in reasonable agreement with prior studies. For example, two earlier studies used a single-muscle model and predicted compressive forces of 145-207% (Cappozzo, 1984) and 100-250% (McGill, 1992) of body weight (BW) during normal walking. In a similar study using an EMG-driven model, Khoo et al. (1995) estimated peak compressive forces ranging between 92% and 345% of BW. Predicted peak compressive forces during normal walking here were similar, with a mean of 187% of BW across all participants. Regarding anterior/posterior L5/S1 shear forces, Khoo et al. (1995)reported a mean equal to 22% BW during gait (Khoo et al., 1995,) while Callaghan et al. (1999) estimated it to be 15%. Anterior/posterior shear forces under the same condition here were 37% of BW across all participants. The somewhat larger values found here might stem from methodological differences in obtaining the L5/S1 joint center of rotation, experiment sample size, participant gender, or walking speed. Moreover, lateral shear was estimated as 12-58% of BW by Callaghan et al. (1999) during normal walking, which is compatible with a mean value of 12% predicted here. Of note, the observed patterns of lumbar angles during normal walking in this investigation are also qualitatively and quantitatively consistent with earlier reports (Cheng et al., 1998; Rowe & White, 1996).

Earlier work has highlighted back pain as a potential outcome of slip and trip incidents (Grönqvist et al., 2001; Manning & Shannon, 1981a; Manning et al., 1984; Pope, 1989; Rohrlich et al., 2014; Troup et al., 1981). More generally, large spinal loads (compression and shear) are among the important biomechanical indicators suspected in the onset of low back disorders, especially in occupational settings (Marras, 2000; McGill, 2015). Peak L5/S1 compression forces during both slip and trip events found here were ~ 2800 N on average. However, peak compression forces estimated from four participants (~33%) exceeded the "Action Limit" of 3400 N by NIOSH (1981). More generally, we estimated that both genders had peak compression forces on the order of 2.2 to 3.9 kN during both slip and trip recoveries. The mean age in the current sample was 24 years, and the upper limit of the noted compression magnitudes are slightly **below** the age-specific lower limits of compression tolerance as reported by Jäger (2018). Despite males being taller and having higher body mass, we found no significant differences between genders in spine kinematics and kinetics. Given a lower average lumbar compression tolerance (Jäger, 2018), the latter suggests that injury risk may be higher among females during both slip and trip recoveries. If one assumes that similar peak compression forces would be observed during slip and trip recoveries among older individuals, their lower average compression tolerance would also suggest higher injury risks.

We estimated peak shear forces that were > 800 N for both perturbations (i.e., slip and trip), larger than the "Action Limit" of 500 N proposed by McGill et al. (1998). More recently, a shear action limit of 1000 N was suggested by Gallagher and Marras (2012), particularly when loading is not repetitive (such as during an unexpected slip or trip). Peak shear forces during slip and trip perturbations were > 500 N for all participants, and > 1000 N for five of them (~40%). Of note, comparing our findings reported herein to our earlier study (Rashedi et al., 2012), both compression and shear forces were estimated to be somewhat smaller, likely due to protocol-related differences in conducting the experiment, including the material used to make the surface slippery.

In addition to high L5/S1 forces, the initial phases of slip and trip recoveries involved rather high levels of muscle activity (Table A4), which is comparable with results provided by Mueller et al. (2016). Specifically, these authors reported that muscle activity in the lumbar region increased 1.8-11 fold under different perturbation conditions imposed by a treadmill, compared to normal walking. Similarly, we found a 2-12 fold increase in lumbar muscle activity during slip or trip recoveries compared to normal walking. Notably, mean values of peak muscle activations were roughly comparable, on the order of 65-75% MVC, though some reached maximum activation levels during slip/trip recovery and there was substantial variability between individuals. These high levels of activity were found across all muscles monitored (Table A4) and included considerable levels of antagonistic co-contraction. Such antagonistic co-contraction can play a key role in augmenting stability of the spine (Gardner-Morse & Stokes, 1998). Hence, we posit that in response to unexpected dynamic perturbations such as slip and trip events, the neuromuscular system uses reflexive “stiffening” of the trunk in the early phases of recovery. However, this co-contraction leads to high levels of low-back loads, which in turn could also contribute to musculotendinous damage and the initiation or progression of low-back disorders.

Two important limitations in our current study need to be acknowledged. The first limitation pertains to problems experienced with force platform data. We were not confident in the quality of this data from reviewing it during initial analyses (e.g., we found that signals were saturated in perturbation trials across several participants). Thus, we could not complete analysis towards evaluated the model-based estimates of lumbosacral moments. As noted earlier, though, there was a level of consensual validity, since model-based estimates during the normal walking condition were consistent with earlier reports. The second is that the participants were told to anticipate gait perturbations (as documented in the informed consent form), which could have predisposed them to muscle tensioning or other physiological changes that might not be present in a non-experimental setting. Nonetheless, we used methods to minimize expectancy effects, by utilizing distracting and secondary tasks during the walking trials. During data collection, we also checked head orientation and visual focus, to the extent possible (e.g., to determine if participants were actively checking the floor for any slippery surfaces). If such activities were observed, we terminated the session and excluded their data from our analysis **(which occurred in approximately two cases)**. Given this, it is reasonable to anticipate more substantial kinematic/kinetic deviations from normal walking for completely unexpected slip and trip perturbations than were observed here. Future work is needed, though, to confirm this anticipation. Future studies are also needed to assess the impacts of load carriage during slip and trip responses, which may also lead to larger kinematic/kinetic deviations.

In summary, the results of our study suggest that, in addition to potential causes *subsequent* to a slip or trip event (e.g., contact with the ground from a fall), a low-back injury could result from high forces generated *during recovery* from the perturbation. The findings described herein help to quantify differences in low-back loads between normal and perturbed gait (i.e., during slip or trip), which can then facilitate future identification of potential contributors to low-back injuries such as overexertion. Future work is recommended to determine if there are direct associations between spine loads and either slip/trip severity or the adequacy of the recovery response. Future work could also be undertaken to develop and test effective interventions (e.g., an occupational trip or slip training system) and to assist in the identification of parameters that will further optimize the effectiveness of such strategies (training frequency, duration, intensity, etc.). Doing so could lead to better control of specific mechanisms involved with low-back disorders consequent to a slip or trip.

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Figure 1. A participant experiencing a slip on a lubricated surface (left) and a trip from a "trap-door" mechanism activated by the experimenter (middle). The slippery surface and the trip plate are shown on the right.

Figure 2. Representative trial for one participant, showing lumbar angles – axial rotation (AR), lateral bending (LB), and flexion/extension (FE) – for (a) normal walking, (b) slip, and (c) trip trials. Time = 0 indicates heel contact. Positive angles indicate flexion, right lateral bending, and clockwise axial rotation.

Figure 3. Peak lumbar angles (top), moments (middle), and reaction forces (bottom) during normal walking, slip, and trip trials. Errors bars are standard deviations. For all results illustrated, **differences between slip and trip outcomes were not statistically different, yet both outcomes were** significantly different than during walking.

Figure 4. Normalized EMG (nEMG) of the lumbar muscles for one participant during: (a) normal walking, (b) slip, and (c) trip trials. Time = 0 indicates right heel contact.

Figure 5. Means of maximum normalized muscle activations for the 14 lumbar muscles during normal walking, slip, and trip trials. Error bars are standard deviations.

Figure **6**. Lumbar kinetics for one participant. Triaxial L5/S1 reaction moments (top) and forces (bottom) for: (a) normal walking, (b) slip, and (c) trip trials. Right foot heel contact is at time = 0. Positive forces indicate compression, anterior and right lateral directions.

Figure 1

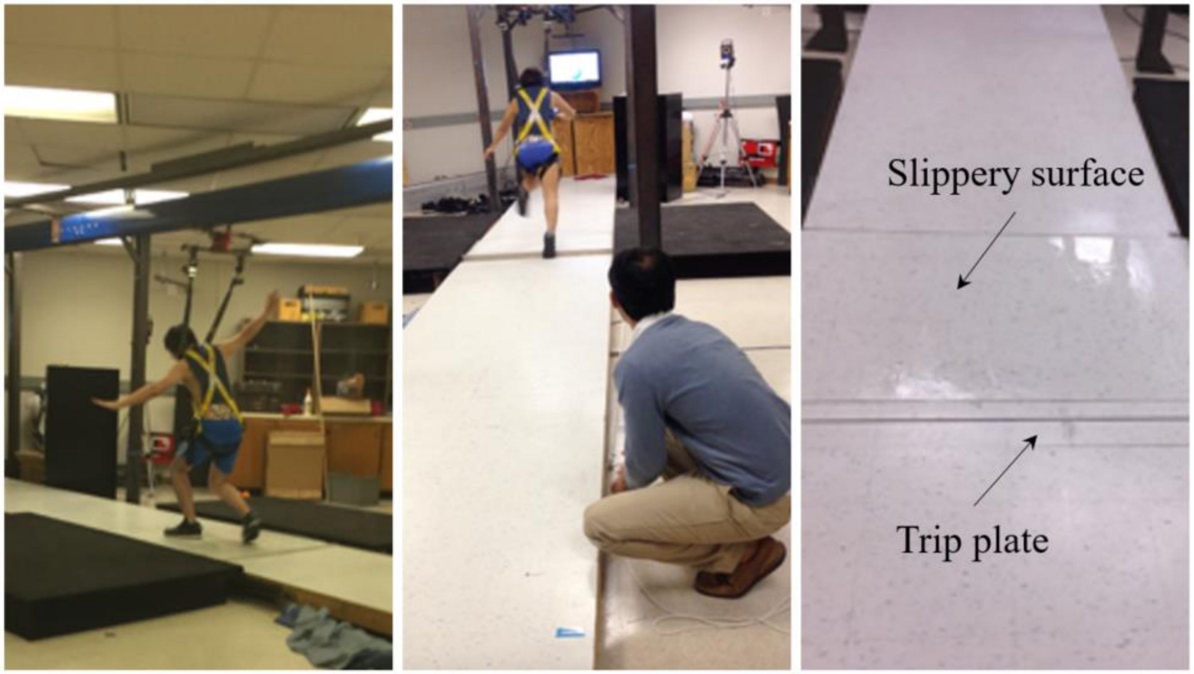
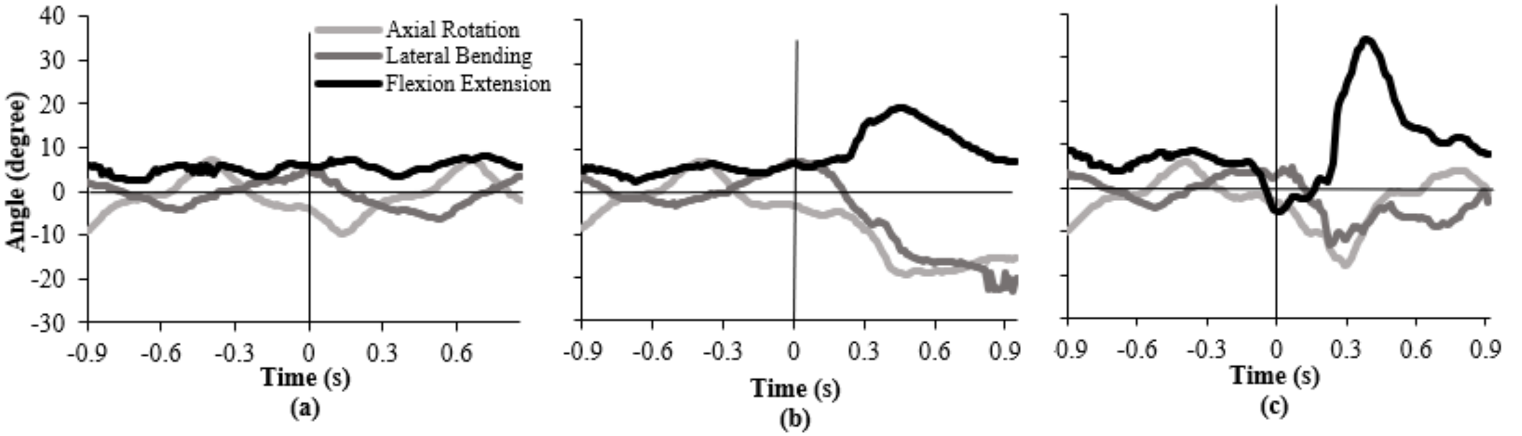


Figure 2



40
30
20
10
0
-10
-20
-30
Angle (degree)
Axial Rotation
Lateral Bending
Flexion Extension
-0.9
-0.6
-0.3
0
0.3
0.6
0.9
Time (s)
(a)
(b)
(c)

Figure 3

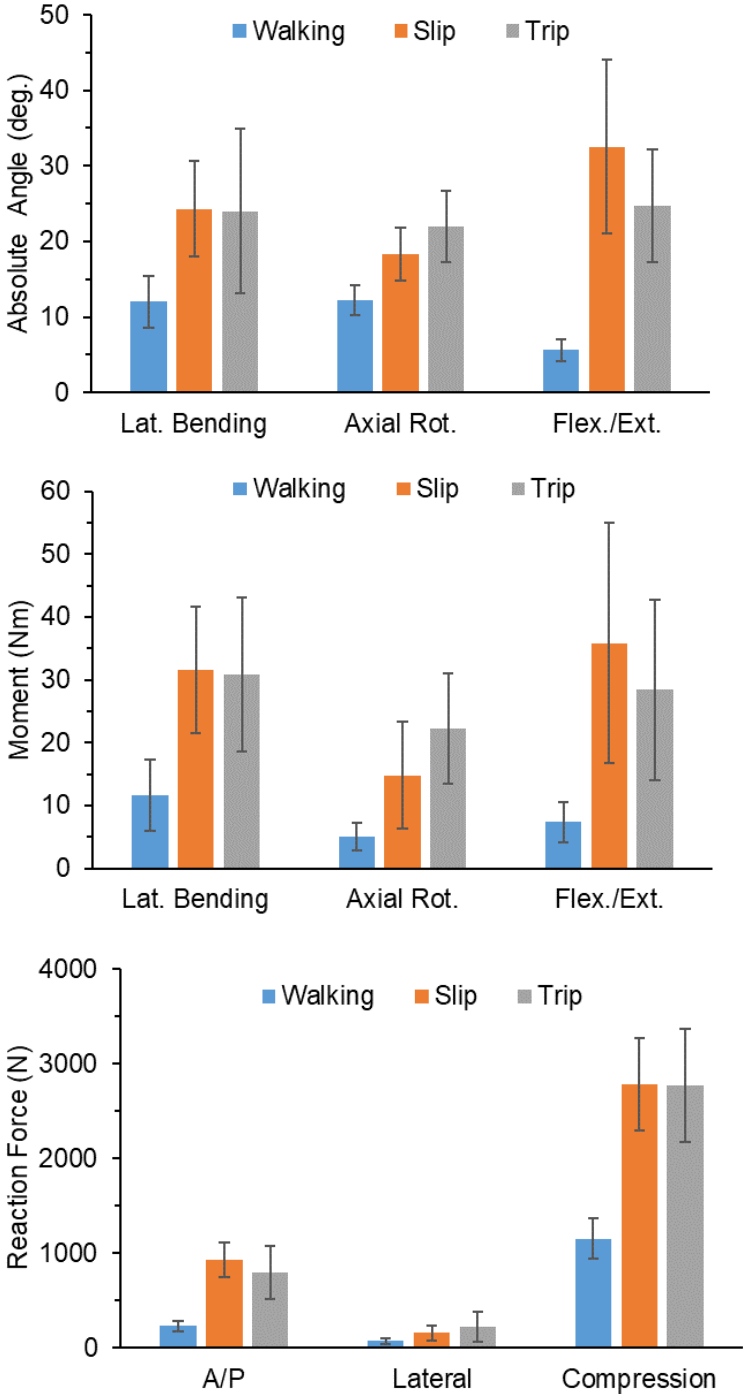
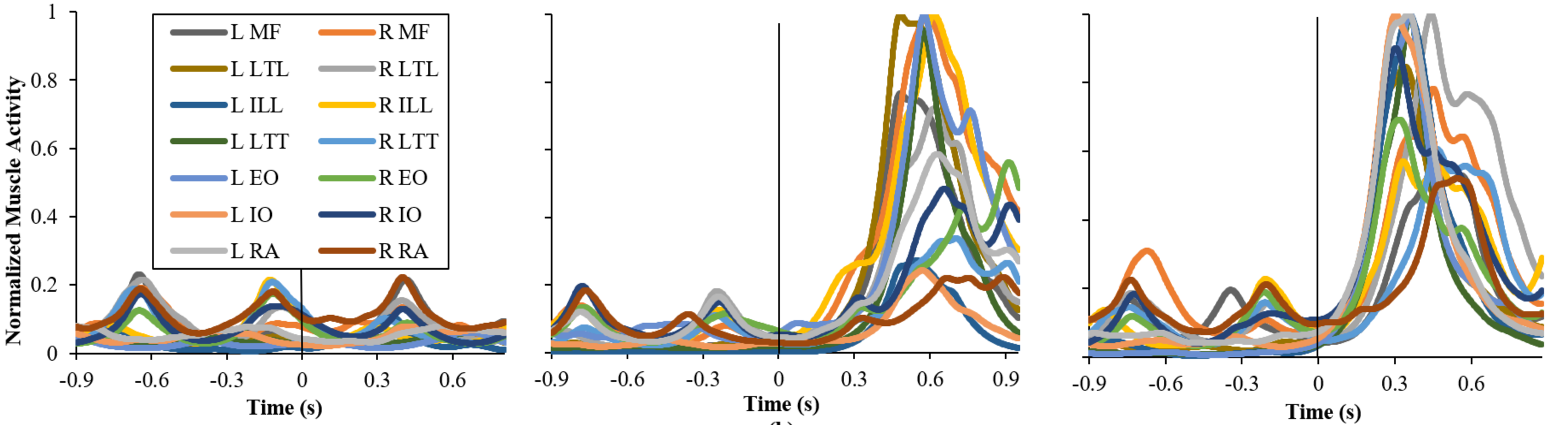

50
40
30
20
10
0
Absolute Angle (deg.)
Walking
Slip
Trip
Lat. Bending
Axial Rot.
Flex./Ext.
60
50
40
30
20
10
0
Moment (Nm)
Walking
Slip
Trip
Lat. Bending
Axial Rot.
Flex./Ext.
4000
3000
2000
1000
0
Reaction Force (N)
Walking
Slip
Trip
A/P
Lateral
Compression

Figure 4

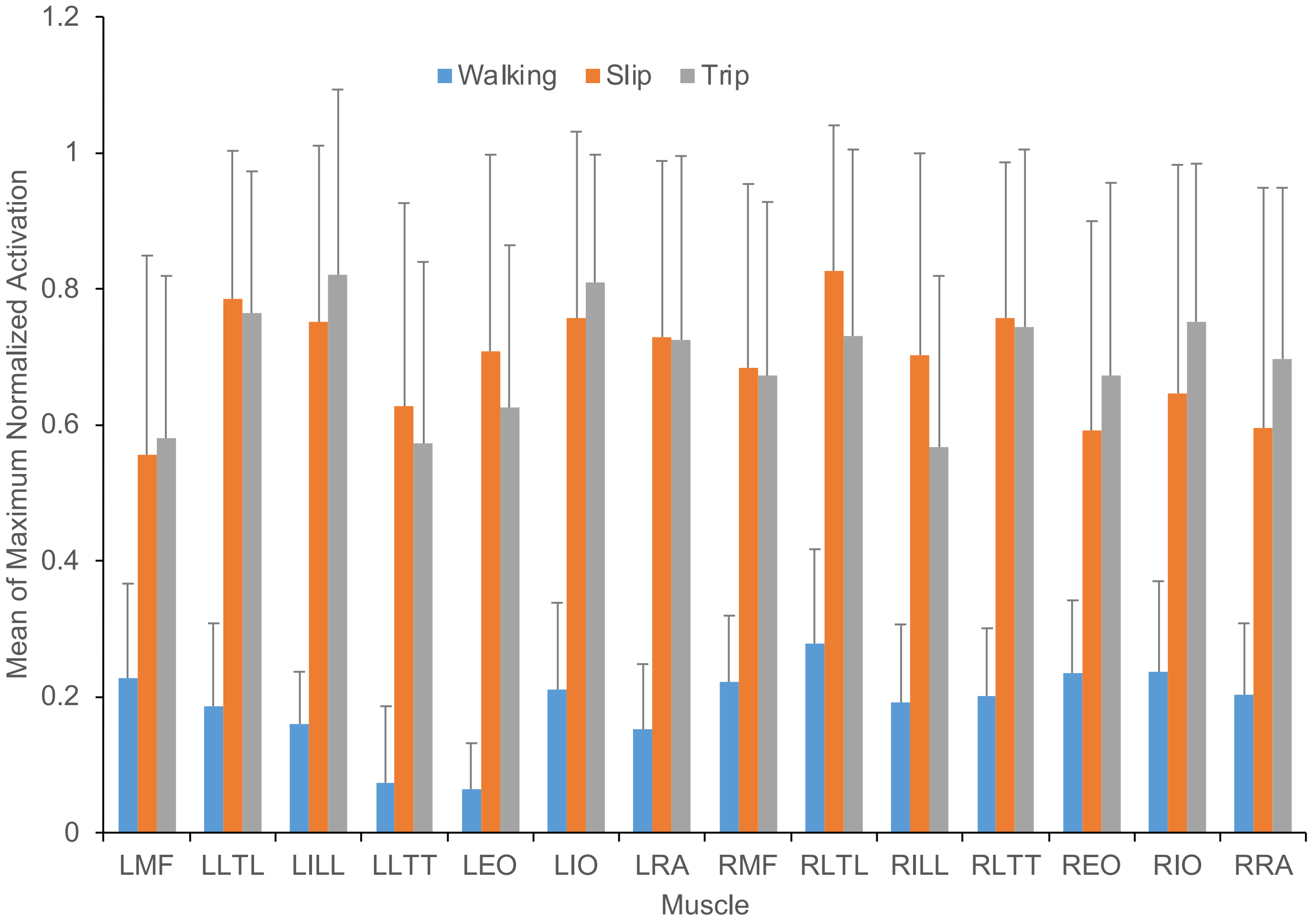

Normalized Muscle Activity
1
0.8
0.6
0.4
0.2
0
L MF
R MF
L LTL
R LTL
L ILL
R ILL
L LTT
R LTT
L EO
R EO
L IO
R IO
L RA
R RA
-0.9
-0.6
-0.3
0
0.3
0.6
0.9
Time (s)

(a)

(b)

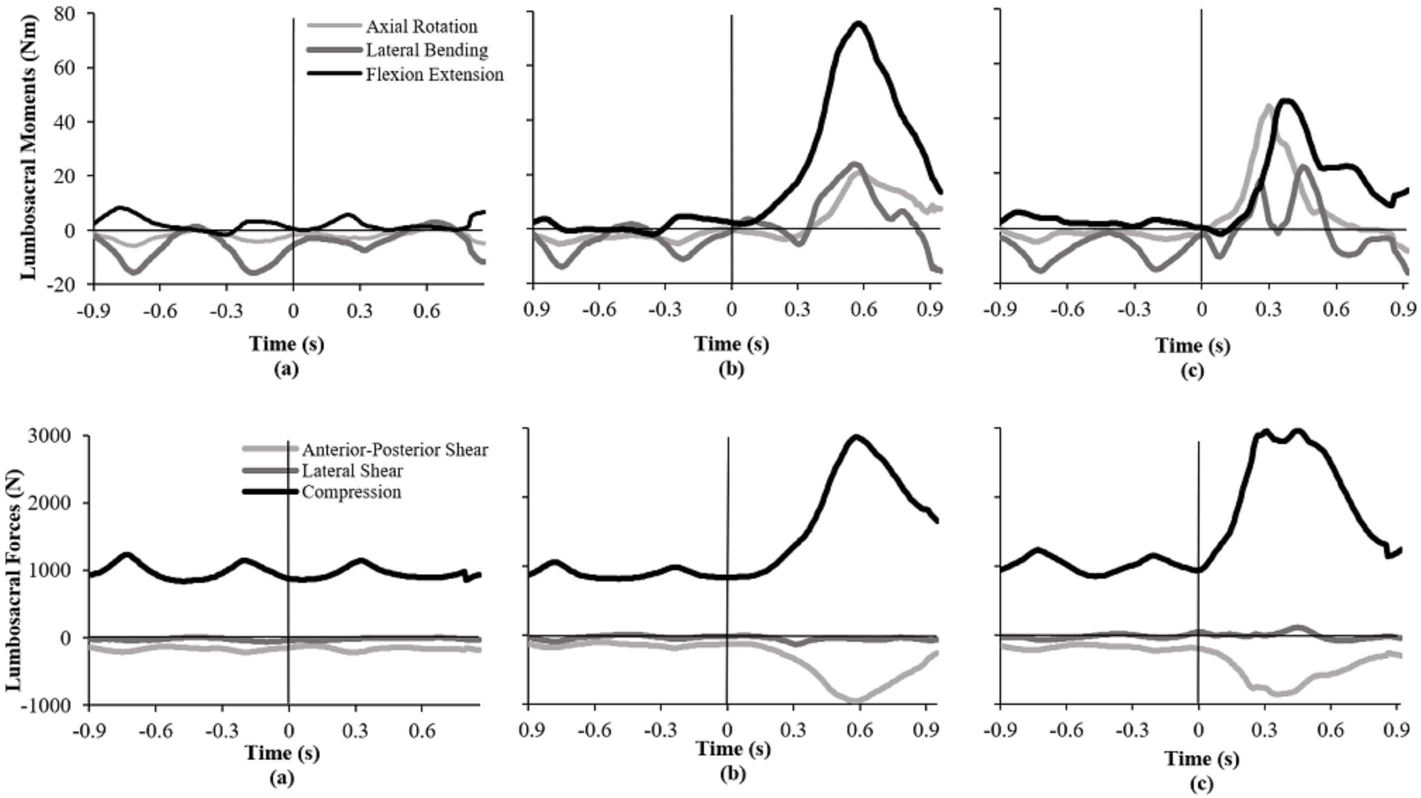
(c)

Figure 5



Walking
Slip
Trip
Mean of Maximum Normalized Activation
1.2
1
0.8
0.6
0.4
0.2
0
LMF
LLTL
LILL
LLTT
LEO
LIO
LRA
RMF
RLTL
RILL
RLTT
REO
RIO
RRA
Muscle

Figure 6



Lumbosacral Moments (Nm)
80
60
40
20
0
-20
Axial Rotation
Lateral Bending
Flexion Extension
-0.9
-0.6
-0.3
0
0.3
0.6
0.9
Time (s)
(a)
(b)
(c)
Lumbosacral Forces (N)
3000
2000
1000
0
-1000
Anterior-Posterior Shear
Lateral Shear
Compression