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Original Research

Effect of Jump Direction and External Load on Single-Legged Jump-Landing Biomechanics

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ABSTRACT

International Journal of Exercise Science 13(1): 234-248, 2020. External load may increase an individual's risk of non-contact anterior cruciate ligament (ACL) injury during single-legged jump-landing (SLJL). This study evaluated the effects of jump direction and external load on hip and knee joint motion and time to stabilization (TTS) during SLJL. Seventeen active males (n = 8) and females (22.2 ± 3.0 y, 1.75 ± 0.08 m, 73.4 ± 12.0 kg) participated in this randomized, crossover designed study. Single-legged jump-landings performed in two conditions, including without external load (BW) and with a torso-worn weight vest equal to 10% of the participant's body weight (BW+10%), from backward, forward, medial, and lateral SLJL directions. Two-way repeated measures ANOVA did not identify any significant interactions ($P > .01$, η^2: < .001 - .037), but some main effects for condition with small effect sizes were identified ($P < .01$, η^2: .009 - .039). Several main effects for SLJL direction were identified with larger effect sizes ($P < .01$, η^2: .010 - .574). This suggests SLJL direction may challenge different components of SLJL biomechanics, and that recreationally active, college-aged individuals may possess effective compensatory mechanisms that can mitigate the effect of BW+10%.

KEY WORDS: Weight vest, hopping, anterior cruciate ligament, injury risk

INTRODUCTION

Anterior cruciate ligament (ACL) ruptures may occur during single-legged jump-landings (SLJL). The effect of external load, such as an additional 10% of the individuals' body weight, and jump direction on SLJL biomechanics associated with ACL injury is not well understood. Approximately 50% of all ACL injuries are incurred by individuals between 15 and 25 years of age, often during routine athletics or physical activity (12). Despite the high proportion of patients that achieve normal knee joint function, surgery to restore knee joint integrity and subsequent rehabilitation do not eliminate long-term sequelae (23, 29, 31). Identifying risk factors and those at risk for ACL injury or re-injury using diagnostically accurate tests and appropriately intervening to correct the factors associated with the increased risk is of paramount importance (15, 24).

Of the ACL injuries that occur during sport participation, it has been estimated that approximately 70% of them are a result of a non-contact mechanism during high-risk maneuvers that require the proper attenuation of ground reaction forces (GRF), such as SLJL (5). As some of the injuries arising from these mechanisms are thought to be preventable, a significant amount of attention has been devoted to prospectively identifying risk factors that predict them. Limited sagittal plane hip and knee motion, greater stiffness, and poor dynamic postural stability (DPS) as measured by time to stabilization (TTS), are some potentially modifiable risk factors that have been identified and may be captured by SLJL (5, 9, 14, 21, 22, 30).

Poor attenuation of GRF is often done using minimal knee flexion angles at initial contact and change in knee flexion angles throughout the weight acceptance phase of SLJL, which increases stiffness and may lead to additional aberrant motion in the frontal plane (5, 6, 14, 21, 22, 30). When this occurs during SLJL, the ACL's ability to rapidly provide passive restraint can be directly challenged (25, 26). While the assessment of DPS does not directly evaluate for the presence of adequate attenuation, the kinetic profile displayed during a SLJL may serve as a surrogate measure of the capacity to achieve stability during a shift from a dynamic movement to a stationary position over the base of support (9, 32). Time to stabilization is one objective measure of DPS (9, 32).

As TTS may serve as a surrogate measure of ACL injury risk, the question remains about the effect of external load on TTS. Literature has demonstrated that external load may have an effect on lower-extremity biomechanics during jump-landings, which may increase the risk for musculoskeletal injury (8, 16, 19, 20, 33). External load has frequently been applied to the trunk. This is commonly done to simulate the load and placement of equipment used by military members and first responders (8, 33). Small loads, such as 10% of body weight, may have a detrimental effect on landing biomechanics (16, 19, 20). These effects, in part, may be due to a decreased strength to body weight ratio, greater inertia of the trunk, or a change in the center of mass (COM) of the trunk segment and an associated change in the radius of gyration of the trunk about the pelvis. To put this small additional load in the context of athletics, an American football player's pads and helmet have a weight similar to 10% of the player's body weight. Furthermore, training or warming-up with a weighted vest to improve jumping ability or subsequent athletic performance is common practice in athletics (10, 17).

Accordingly, the purpose of this study was to determine the effect of external load equal to 10% of the participant's body weight and SLJL direction on TTS and other biomechanical measures. These measures include change in the sagittal plane knee and hip joint angle during the weight acceptance period, peak vertical COM position during the flight phase, COM vertical velocity at initial contact, and change in whole body COM vertical position during the weight acceptance period. It was hypothesized that SLJL performed with external load would elicit greater TTS as well as hip and knee joint motions associated with biomechanics that put an individual at a greater risk for non-contact ACL injury. Furthermore, it was hypothesized that a backwards SLJL would demonstrate TTS and hip and knee joint motions that were more severe in relation to ACL injury risk compared to a forwards, medial, or lateral SLJL.

METHODS

Participants

Seventeen recreationally active males and females (Females = 9, Males = 8, 22.2 ± 3.0 y, 1.75 ± 0.08 m, 73.4 ± 12.0 kg) participated in this exploratory study. The cohort engaged in their primary modes of physical activity, which included running, biking, resistance training, soccer, basketball, and volleyball amongst others, on average three to five days per week or six to ten hours per week. Participants were excluded if they had a history of ligamentous, bony, or other soft tissue reconstruction to the lower-extremity, an orthopedic issue exacerbated by exercise, or a neurological issue that affected the activation of skeletal muscle or balance. All aspects of this study were approved by the Sanford Health Institutional Review Board. In conjunction with human subjects in research policies, all subjects were informed of the potential risks and benefits of participating in this study before providing their informed, written, voluntary consent. None of the participants were less than 18 years old. This research was carried out fully in accordance to the ethical standards of the International Journal of Exercise Science (28).

Protocol

This exploratory study used a randomized, cross-over, within-subjects design to assess the effect of external load, in the form of a torso worn weight vest that was equal to 10% of the participant's body weight, on hip and knee joint kinematics as well as TTS during SLJL. Participants attended two sessions. During the first session, the participants were adequately familiarized to the SLJL protocol. At the second session, participants completed a series of SLJL on the dominant leg from forward, backward, lateral, and medial directions. Participants completed the series of SLJL twice, once with external load (BW+10%) and once without external load (BW), in a randomized order. The second visit was completed three to ten days after the first visit.

During visit 1, the participant's height, body weight, age, and leg dominance, which was the leg the participant indicated they would kick a soccer ball with, were recorded. Participants then completed a standardized warm-up consisting of light intensity aerobic exercise, dynamic stretching, and plyometrics, which took approximately 10 minutes to complete. Participants were then familiarized to the SLJL protocol, described below, by performing three BW trials followed by three BW+10% trials from the four jump directions.

During visit 2, the participant's body weight was re-measured. Small weights were distributed in a balanced fashion in the pockets of a torso worn weight vest until the weight of the vest equaled 10% of the participant's body weight. This weight was chosen in order to mimic the weight of common athletic protective equipment, such as football pads and helmet. Participants then completed the standardized warm-up that took approximately 10 minutes. They were then re-familiarized to the SLJL protocol, explained below, by completing three BW trials from the four jump directions, which took approximately five minutes to complete.

Retro-reflective markers were placed on the participants by a single investigator in locations associated with the pelvis, thigh, shank, and foot segments using a modified Helen Hayes model. A 12 camera, three-dimensional, infrared motion capture system sampling at 240 Hz and accompanying software (Cortex, Motion Analysis Corporation, Rohnert Park, CA, USA) was used to obtain the trajectories of the markers. Side-by-side 0.6 x 0.9 m, three-dimensional, in-ground force plates (Bertec, Columbus, OH, USA) level with the surrounding floor were used to collect kinetic data during SLJL. The force plates were synchronized with the motion capture system and sampled at 1,200 Hz. For all SLJL trials, all participants jumped off the same force plate (take off force plate) and landed on the same force plate (landing force plate).

The SLJL protocol was similar to the protocol described by DuPrey et al. (2016), and included jumps from a forward, backward, lateral, and medial direction (9). For the forward SLJL, a 0.15 m tall hurdle was placed equidistant to the participant's leg length from the center of the landing force plate parallel to the horizontal axis of the force plates. A line was then drawn on the floor parallel to this axis that was 1.5 times the participant's leg length from the hurdle. Starting from this line, participants took two comfortable steps before jumping over the hurdle on the test leg and landing on the test leg on the landing force plate. For the other jump directions, the hurdle was placed directly next to the landing force plate. Participants jumped over the hurdle on the test leg and landed on the test leg on the landing force plate. For all jump directions, participants were given an audible signal from the investigator to initiate their SLJL. Participants were instructed to land with their eyes focused forward with their hands placed on their hips, stabilize as quickly as possible, and remain as motionless as possible until the investigator indicated that the trial was over. A minimum of 10 seconds of kinematic and kinetic data were obtained. Jumps were performed barefoot to minimize the stability provided by a shoe.

After the conclusion of the refamiliarization trials, approximately 10 minutes were taken to place the retroreflective markers and capture the necessary calibrations for post-hoc modeling. The order of the directions (e.g. forward, backward, medial, lateral) and conditions (e.g. BW vs BW+10%) completed were randomized to avoid an order effect. Three trials per direction per condition were completed, totaling 24 trials. Trials in which the participant stumbled, the non-test leg made contact with the floor, or hands were removed from the hips were discarded and an additional trial was completed. After completion of the first condition, participants rested stationary for 15 minutes before starting the second condition.

Key phases of a representative backward SLJL are depicted in Figure 1. Ground reaction force and marker trajectory data were filtered post hoc using a 2nd order low-band pass 12 Hz and 6 Hz butterworth filter, respectively. The pelvis origin of the model was used as a surrogate position for the participant's whole body COM. Peak vertical position of COM between SLJL take-off and landing (e.g. initial contact), which represents the flight phase timeframe, was recorded. The weight acceptance phase of the SLJL was defined as the instant when vertical ground reaction (vGRF) force first exceeded 20 N and the first peak vGRF following initial contact. The vertical velocity of COM at initial contact was recorded. The change in the sagittal plane knee and hip joint angle (e.g. extension-flexion axis) were calculated as the difference between the joint angle at initial contact and peak vGRF. The change in whole body COM vertical position was also calculated as the difference between initial contact and peak vGRF. TTS was quantified as the time required for the vGRF to reach and remain within ± 5% of the participant's BW or BW+10%, depending upon condition, for one second after initial contact (9).

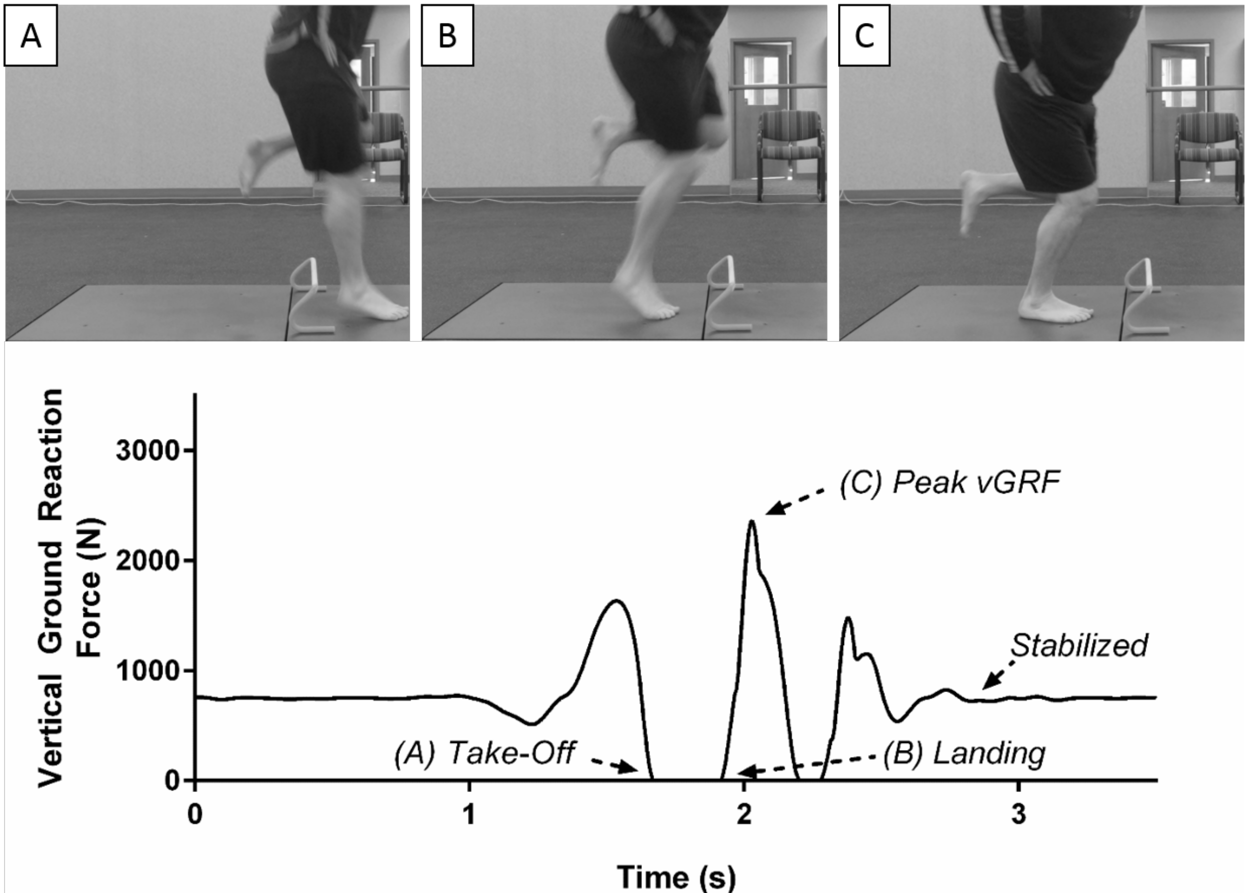

Figure 1. Representative vertical ground reaction force (vGRF) curve for a backward single-legged jump-landing. Curve depicts the phases take-off (A), landing (B), peak vGRF (C), and when stabilization occurs. The flight phase is from take-off to landing (A to B), the weight acceptance period is from landing to peak vGRF (B to C), and time to stabilization is from landing to stabilized. The change in the sagittal plane knee and hip joint angle (e.g. extension-flexion axis) were calculated as the difference between the joint angle at initial contact (B) and peak vGRF (C).

Statistical Analysis

Statistics were completed using Prism (Graph Pad, San Diego, CA, USA). The dependent variables of this study were the change in the sagittal plane knee and hip joint angle during the weight acceptance period, peak vertical COM position during flight phase, vertical velocity of COM at initial contact, change in whole body COM vertical position during the weight acceptance period, and TTS. Separate 2x4 (condition x direction) two-way repeated measures analysis of variance (ANOVA) were used to assess for the effect of external load (e.g. BW vs BW+10%) and jump direction (e.g. backward, forward, lateral, and medial) on the dependent variables. The ANOVAs were Geisser-Greenhouse corrected to account for potential violations of the sphericity assumption. If appropriate, Tukey corrected multiple comparisons were used to identify the source of the effect. Due to the large number of ANOVAs completed, alpha was initially set to $P \leq .01$ for all statistical tests.

RESULTS

Time to Stabilization: External load and SLJL direction did not have an effect on TTS (Figure 2). Specifically, an ANOVA indicated that there was no significant interaction effect (F (1.96, 31.37) = 2.12, P = .137, η^2 = .037, ε = .653), nor a significant main effect for condition (F (1, 16) = 1.88, P = .189, η^2 = .009, ε = 1) and direction (F (2.80, 44.75) = .376, P = .757, η^2 = .010, ε = .932) on TTS.

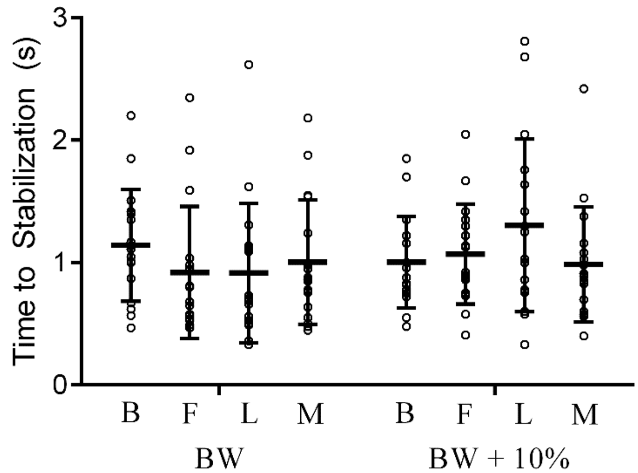

Figure 2. Time to stabilization (s) during backward (B), forward (F), lateral (L), and medial (M) single-legged jump-landings with external load (BW+10%) and without external load (BW). Clear circles: individual participants; whiskers: standard deviation.

Joint Angle Changes: Change in the sagittal plane knee and hip joint angle were calculated as the difference between the joint angle at initial contact and peak vGRF. The magnitude of change may represent the degree of eccentrically controlled hip and knee flexion used to attenuate the SLJL forces. External load and SLJL direction did not have an effect on sagittal plane hip joint kinematics during the weight acceptance period (Figure 3, Table 1). An ANOVA indicated that there was no significant interaction effect (F (2.38, 38.06) = 2.50, P = .086, η^2 = .022, ε = .793), main effect for condition (F (1, 16) = 6.97, P = .018, η^2 = .027, ε = 1), or main effect for direction (F (2.48, 39.70) = 4.60, P = .011, η^2 = .067, ε = .827) on change in sagittal plane hip joint angle during the weight acceptance period.

At the knee joint, external load also did not have an effect on sagittal plane kinematics during the weight acceptance period, but SLJL direction did have an effect. An ANOVA indicated that there was no significant interaction effect (F (2.11, 33.73) = 2.44, P = .100, η^2 = .014, ε = .703) and main effect for condition (F (1, 16) = 5.05, P = .039, η^2 = .011, ε = 1), but there was a significant main effect for direction (F (2.36, 37.71) = 70.18, P < .001, η^2 = .574, ε = .786) on change in sagittal plane knee joint angle during the weight acceptance period.

Table 1. Change in knee joint sagittal plane angle from initial contact to peak vertical ground reaction force (Δ°) mean differences [99% confidence interval of the difference] for the marginal means of single-legged jump-landing direction. Table presents only the marginal means comparisons for the effect of direction that were significant (P ≤ 0.01).

Direction (Mean ± SD)	P-value	Mean Diff. [99% CI]
Change in Knee Sagittal Plane Angle (Δ°)		
Forward (23.6 ± 0.7) > Backward (-2.6 ± 4.8)	< .001	26.2° [19.5, 32.9]
Lateral (10.9 ± 2.4) > Backward (-2.6 ± 4.8)	< .001	13.5° [8.8, 18.1]
Medial (12.3 ± 1.0) > Backward (-2.6 ± 4.8)	< .001	14.8° [8.6, 21.0]
Forward (23.6 ± 0.7) > Lateral (10.9 ± 2.4)	< .001	12.7° [6.8, 18.7]
Forward (23.6 ± 0.7) > Medial (12.3 ± 1.0)	< .001	11.4° [5.3, 17.5]

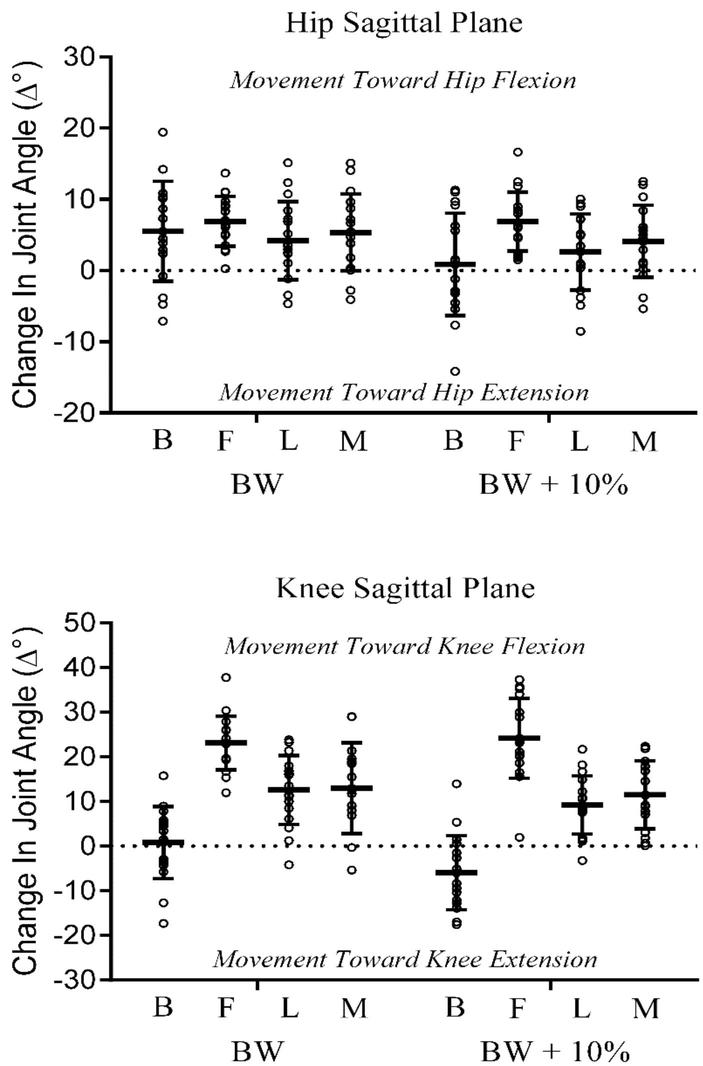

Figure 3. Change in hip and sagittal plane joint angle from initial contact to peak vertical ground reaction force (Δ°) backward (B), forward (F), lateral (L), and medial (M) single-legged jump-landings with external load (BW+10%) and without external load (BW). Clear circles: individual participants; whiskers: standard deviation.

Center of Mass: External load did have effect on peak vertical COM position during the flight phase and vertical change in COM position during the weight acceptance period, but not on the COM downward vertical velocity at initial contact (Figure 4, Table 2). Single-legged jump-landing direction had an effect on all three COM variables. An ANOVA indicated that there was no significant interaction effect (F (2.77, 44.38) = .3687, P = .761, $\eta^2 < .001$, ε = .925), but there was a significant main effect for condition (F (1, 16) = 15.99, P = .001, η^2 = .013, ε = 1) and direction (F (1.985, 31.65) = 6.23, $P < .005$, η^2 = .025, ε = .659) on peak vertical COM position during the flight phase. The participants COM achieved a peak vertical position of 1.153 ± .010 m during BW and 1.140 ± .011 m during BW+10% (mean difference = .013 m, 99% CI [.003, .022]). An ANOVA also revealed that there was no significant interaction effect (F (2.30, 36.85) = .370, P = .722, η^2 = .002, ε = .768) and main effect for condition (F (1, 16) = 2.94, P = .106, η^2 = .011, ε = 1), but there was a significant main effect for direction (F (2.15, 34.31) = 53.81, $P < .001$, η^2 = .323, ε = .715) on COM vertical velocity at initial contact. Finally, an ANOVA indicated that there was no significant interaction effect (F (2.24, 35.85) = .180, P = .859, η^2 = .002, ε = .747), but there was a significant main effect for condition (F (1, 16) = 11.31, P = .004, η^2 = .037, ε = 1) and direction (F (2.46, 39.29) = 17.92, $P < .001$, η^2 = .258, ε = .818) on change in whole body COM vertical position during the weight acceptance period. The participants change whole body COM vertical position was .093 ± .016 m during BW and .081 ± .018 m during BW+10% during the weight acceptance period (mean difference = .012 m, 99% CI [-.004, .028]).

Table 2. Peak center of mass height achieved during the jump (m), center of mass vertical velocity at initial contact ($m \cdot s^{-1}$), and change in center of mass position from initial contact to peak vertical ground reaction force (Δm) mean differences [99% confidence interval of the difference] for the marginal means of single-legged jump-landing direction. Table presents only the marginal means comparisons for the effect of direction that were significant ($P \leq 0.01$).

Direction (Mean ± SD)	P-value	Mean Diff. [99% CI]
Peak Center of Mass Height (m)		
Forward (1.161 ± .007) > Backward (1.139 ± .007)	.007	.022 [< .001, .043]
Forward (1.161 ± .007) > Medial (1.146 ± .012)	.001	.021 [.003, .039]
Forward (1.161 ± .007) > Lateral (1.140 ± .008)	.008	.016 [< .001, .031]
Center of Mass Velocity At Initial Contact ($m \cdot s^{-1}$)		
Forward (-1.71 ± .04) > Backward (-1.39 ± .02)	<.001	.322 [.201, .439]
Forward (-1.71 ± .04) > Lateral (-1.38 ± .03)	<.001	.333 [.246, .420]
Forward (-1.71 ± .04) > Medial (-1.47 ± .05)	<.001	.245 [.158, .331]
Medial (-1.47 ± .05) > Lateral (-1.38 ± .03)	.002	.088 [.014, .163]
Change in Center of Mass Position (Δm)		
Backward (-.100 ± .004) > Forward (-.062 ± .010)	< .001	.037 [.017, .058]
Lateral (-.091 ± .009) > Forward (-.062 ± .010)	< .001	.029 [.009, .049]
Medial (-.095 ± .008) > Forward (-.062 ± .010)	< .001	.033 [.014, .052]

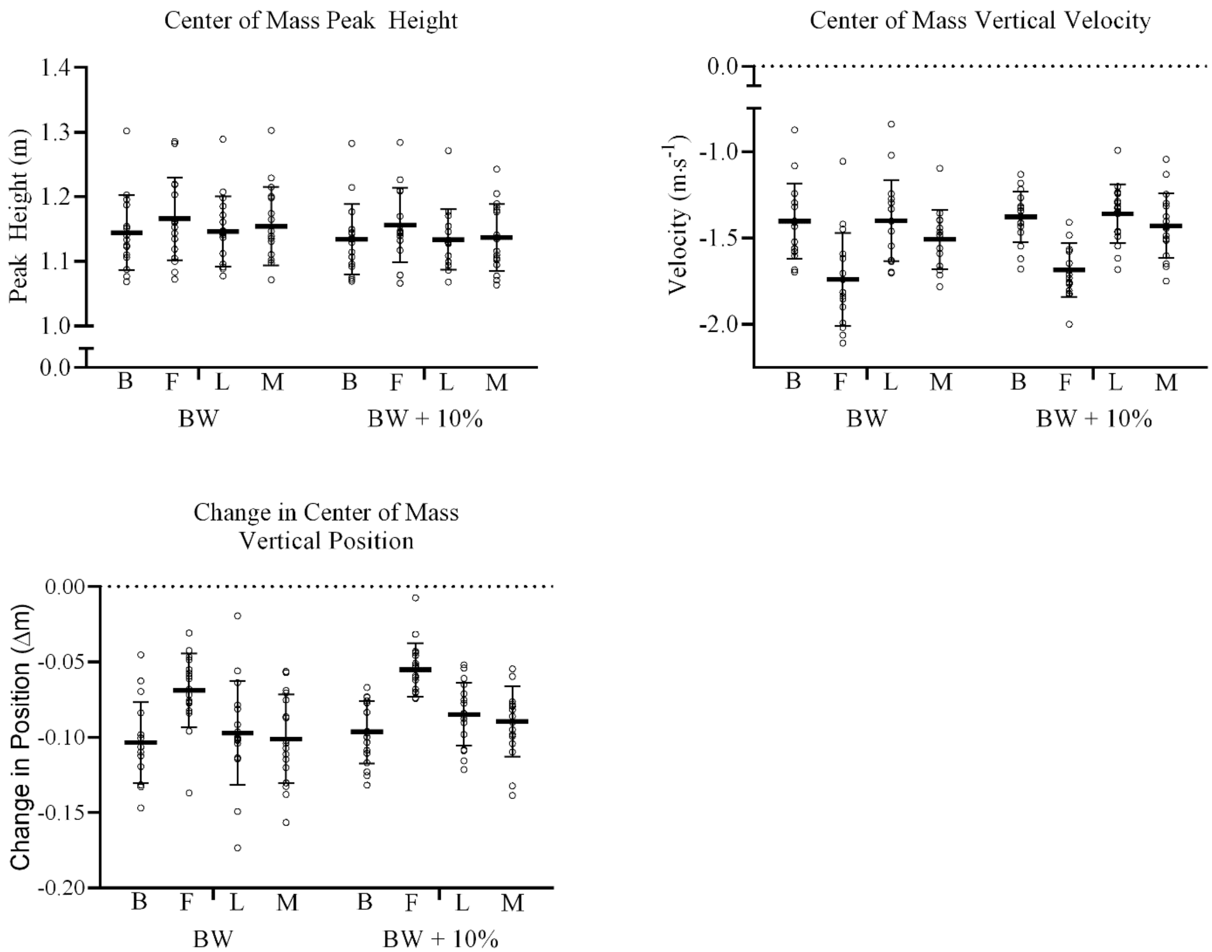

Figure 4. Peak center of mass height achieved during the jump (m), center of mass vertical velocity at initial contact ($m \cdot s^{-1}$), and change in center of mass position from initial contact to peak vertical ground reaction force (Δm) during backward (B), forward (F), lateral (L), and medial (M) single-legged jump-landings with external load (BW+10%) and without external load (BW). Clear circles: individual participants; whiskers: standard deviation.

DISCUSSION

It is estimated that 70% of ACL injuries are the result of a non-contact mechanism (5). Some of these injuries are thought to be preventable through proper identification of predisposing risk factors and targeted intervention, such as an ACL injury prevention program (15, 34). In fact, it has been reported that those participating in an ACL injury prevention program may experience nearly a 50% reduction in ACL injury risk. (36) Past studies have revealed that limited sagittal plane hip and knee motion, greater stiffness, and poor dynamic postural stability (DPS) as measured by time to stabilization (TTS) are some potentially modifiable risk factors. However, the effect of external load and SLJL direction on these factors is not well understood. The purpose of this study was to determine the effect of external load equal to 10% of the participant's body weight and SLJL direction on change in the sagittal plane knee and hip joint angle during the weight acceptance period, peak vertical COM position during the flight phase, COM vertical velocity at initial contact, change in whole body COM vertical position during the weight acceptance period, and TTS. It was hypothesized that SLJL performed with external load would elicit greater TTS as well as hip and knee joint motions associated with biomechanics that put an individual at a greater risk for non-contact ACL injury , and that a backwards SLJL would demonstrate TTS and hip and knee joint motions that were more severe in relation to ACL injury risk compared to a forwards, medial, or lateral SLJL. An important finding from the present study is that the SLJL direction appears to have a large influence on SLJL biomechanics, while a load equal to 10% of body weight does not.

No effect of condition or direction on TTS was identified in the present study, suggesting that the capacity to achieve stability during a shift from a dynamic movement to a stationary position over the base of support was not altered by external load or SLJL direction. The mean TTS values for both BW and BW+10% are comparable to the findings of DuPrey et al. (2016) who prospectively demonstrated that athletes who did not sustain a non-contact ACL injury had similar TTS that was approximately one second in duration from the same four SLJL directions at baseline without external load (9). Although external load is often necessary in the military and athletic sectors, literature has demonstrated that these loads may have a detrimental effect on lower-extremity biomechanics during jump-landings (8, 16, 19, 20, 33). Kulas et al. (2008) demonstrated that BW+10% increased the demands on the knee and ankle joints, potentially increasing the risk for associated injury (20). Findings from the present study, however, suggest that effective compensatory mechanisms likely exist in recreationally active, college-aged adults that can mitigate the effects of BW+10%. This is in line with Janssen et al. (2012), who revealed that BW+10% had no effect on jump-landing biomechanics in highly trained male volleyball players (16).

There are a number of factors associated with neuromuscular control during SLJL which include age, strength, training status, injury status, fatigue, cognitive load, and jump height, that may alter TTS performance (1-4, 7, 8, 13). It is possible that loads greater than BW+10% may be necessary to affect a relatively older or trained individual's neuromuscular control during SLJL, while loads less than BW+10% may affect younger or untrained individuals. This may, in part, be due to insufficient lower-extremity strength of younger and untrained individuals compared to those that are older and trained (1, 13). Furthermore, an individual's injury history may cause deficits in neuromuscular control. Webster and Gribble (2010) demonstrated that female athletes at an average of 2.5 years after ACL reconstruction took approximately .20 seconds longer to stabilize compared to their non-injured peers (35). It is possible that BW+10% may worsen this disparity, though it was not observed in the present study as those with a history of lower extremity traumatic injury were excluded. It is also conceivable that with this relatively low magnitude of external load, factors such as cognitive load and fatigue must be present before significant changes in lower-extremity biomechanics are observed (2, 3, 7). Dempsey et al. (2014) observed small differences in GRF when individuals had external load compared to no external load, and that these differences were enlarged when participants were fatigued and distracted during the landing (8).

It has also been demonstrated that jump height may play an important role in landing biomechanics (4), and that athletes may achieve lower jump heights while externally loaded (16, 18). This is likely because an identical vertical GRF impulse would cause a smaller vertical momentum at take-off due to the added mass of the external load. Ignoring jump direction, participants in the present study achieved peak vertical COM positions approximately 0.01 m higher during the flight phase of BW than they did during BW+10%. This may have washed out any detrimental effect of the external load on TTS, though the rather small effect size (η^2 = .013) would suggest otherwise. Participants routinely achieved greater peak vertical COM positions during the flight phase of the forward SLJL compared to the other three directions. This may be because jumping in this direction is more natural for the majority of people.

Center of mass vertical velocity at initial contact is also, in part, reflective of the jump height achieved as well as other SLJL biomechanics. No effect of condition was identified, indicating that participants had a similar downward velocity at initial contact for both BW and BW+10%, which is generally expected given the aforementioned peak vertical COM position. As expected, the forward SLJL tended to demonstrate a greater COM vertical velocity nearly 0.3 $m \cdot s^{-1}$ faster than the other three directions at initial contact. This suggests that a greater vertical GRF impulse would be required to safely shift from a dynamic movement to a static position during a forward SLJL. One surrogate measure for SLJL stiffness is the change in vertical COM position during weight acceptance, with smaller changes synonymous with stiffer landings (6, 22). An effect of external load was identified for the change in vertical COM position as participants had approximately a 0.01 m smaller change in position during BW+10% compared to BW. This potentially indicates that the BW+10% resulted in stiffer landings which may increase the risk for injury (6, 22). While it may be expected that a greater change in vertical COM position during weight acceptance would be evident during the forward SLJL compared to the other directions due to the increase in vertical velocity at initial contact, the opposite was revealed. Participants had approximately a 0.03 m smaller change in position during the forwards SLJL compared to the other three directions.

To accomplish a greater change in vertical COM position and reduce stiffness during weight acceptance of a SLJL, the participants would have to achieve greater degrees of eccentrically controlled hip and knee flexion, which is discussed in detail below. Increasing the hip and knee flexion angles would likely require an increase in internal hip and knee extension moments, resulting in a greater demand for hip and knee extensor neuromuscular control. Perhaps to avoid this demand participants limited their hip and knee flexion, resulting in a stiffer landing. This may be further exacerbated in individuals with deficits, such as fatigue, cognitive load, or injury history because the difference between this demand and their neuromuscular control capacity has been reduced. Although this stiffer landing strategy may increase the risk for traumatic musculoskeletal injury for these predisposed individuals (21,22,30). Clinicians working with such individuals should emphasize strength development and safe landing strategies, which is similar to the emphasis of many ACL injury prevention programs (5, 34).

While external load did not affect the change in joint angle during weight acceptance in this sample of college aged individuals, the direction in which the SLJL was performed did affect the change in knee joint angle. This suggests that different jump directions may prompt different lower extremity motor control strategies to be used to safely control motion. The safe attenuation of GRF during SLJL is achieved through eccentrically controlled knee and hip flexion. A substantial effect ($\eta^2 = .574$) of SLJL direction was demonstrated for change in sagittal plane knee angle. A slight change in angle corresponding to approximately 2° extension was consistently demonstrated during the weight acceptance period of the backward SLJL. Forward, lateral, and medial SLJL all demonstrated changes corresponding to flexion of 24°, 11°, and 12°, respectively. Given the small change in vertical COM position during weight acceptance for the forward jump, one may expect that the change in sagittal plane knee joint angle would be relatively close to zero in comparison to the other directions. It is possible that attenuation of GRF occurred in more than just the vertical plane for the forward SLJL, as participant's likely had greater horizontal momentum at take-off compared to the backward, lateral, and medial SLJL because they took two initial steps compared to none in accordance with a previous protocol (9).

The change in sagittal plane knee joint angle towards extension during weight acceptance was not expected. It is possible that the backward SLJL challenges an individual's proprioception and visuospatial capabilities as the landing area is not as readily identified during the execution of task, leading to extension at landing compared to other jump directions. This would allow the individual to make contact with the ground faster, which may subconsciously be a preferred strategy. Another explanation is that at landing the COM would be traveling horizontally in a posterior direction. To decelerate the COM an anteriorly directed GRF must be applied, which would require relative extension at the knee in order to prevent the individual from falling over backwards. Simply, the participants had to make contact with ground with their foot posterior to their COM, which resulted in a tendency to extend at the knee. Nevertheless, these challenges may have contributed to why Duprey et al. (2016) demonstrated that TTS during a backwards SLJL was predictive of non-contact ACL injury while other SLJL directions were not (9). An exaggerated backwards SLJL, as performed here, is likely not being performed to the same degree in sport; however, there may be some instances in which the athlete's COM may be moving in a posterior direction, such as rebounding a basketball or performing a block in volleyball. This may yield a similar situation in which they cannot visually identify the landing during the execution of the task with the added complexity of a secondary task, potentially resulting in similar changes in sagittal plane knee joint angle during weight acceptance. To this effect, it has been demonstrated that an athlete may alter their sagittal plane loading strategy when they allocate their attention to secondary tasks (2, 7, 27).

The findings from this study suggest that BW+10% may not affect college-aged individual's TTS and sagittal plane hip and knee joint angle changes during SLJL from a variety of directions, but that SLJL direction has significant effects on a number of important biomechanical measures. These results are of value for sports injury prevention, as it suggests that recreationally active, college-aged individuals are able to effectively compensate for BW+10% and that, from a biomechanical non-contact ACL injury standpoint, external load, such as protective equipment used in American Football or the gear worn by military service members, may be without detriment. Although, as discussed, it is possible that with the accumulation of additional risk factors such as fatigue, injury history, and cognitive load or allocation, a theoretical threshold is reached in which BW+10% may become detrimental. These results also suggest that in order to fully elucidate an individual's injury risk, it may be of value to assess SLJL biomechanics from a variety of jump directions. Each direction may present a unique challenge that leads to the manifestation of aberrant lower-extremity biomechanics when an individual does not have the neuromuscular control capacity to safely execute the SLJL task. For example, clinicians should utilize SLJL from a variety of directions to comprehensively retrain an individual to land safely, and the return to play decision following traumatic injury should consider how the athlete performs from a variety of jump directions rather than just forwards.

This study is not without limitation. Kinematic data from the trunk and upper extremities were not obtained. It is possible that, given the placement of the vest around the upper torso, greater sway of the trunk relative to the pelvis occurred during BW+10%. In this context, poor trunk proprioception and stability has been identified as a risk factor for ACL injury (38). The current study was also limited to 17 individuals, and the homogenous nature of the participants in the present study does not allow for the results to be generalized to all individuals, particularly those that are younger or inactive. TTS can be computed using a range of thresholds, sampling rates, and data filtering techniques, which can limit interpretation of TTS values obtained from different studies (11). TTS only utilizes vGRF. It is possible that other multi-plane measures of DPS may provide a more comprehensive assessment of DPS (37). Finally, joint moments were not assessed in the present study, which may be telling with respect to non-contact ACL injury risk during a SLJL (14)

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