Methodological considerations of task and shoe wear on joint energetics during landing

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Abstract:

To better understand methodological factors that alter landings strategies, we compared sagittal plane joint energetics during the initial landing phase of drop jumps (DJ) vs. drop landings (DL), and when shod vs. barefoot. Surface electromyography, kinematic and kinetic data were obtained on 10 males and 10 females during five consecutive drop landings and five consecutive drop jumps (0.45 m) when shod and when barefoot. Energy absorption was greater in the DJ vs. DL (P = .002), due to increased energy absorption at the hip during the DJ. Joint stiffness/impedance was more affected by shoe condition, where overall stiffness/impedance was greater in shod compared to barefoot conditions (P = .036). Further, hip impedance was greater in shod vs. barefoot for the DL only (via increased peak hip extensor moment in DL), while ankle stiffness was greater in the barefoot vs. shod condition for the DJ only (via decreased joint excursion and increased peak joint moment in DJ vs. DL) (P = .011). DJ and DL place different neuromechanical demands upon the lower extremities, and shoe wear may alter impact forces that modulate stiffness/impedance strategies. The impact of these methodological differences should be considered when comparing landing biomechanics across studies.

Keywords: Landing energetics | Work absorption | Landing biomechanics | Force dissipation strategies | Footwear

Article:

1. INTRODUCTION

Joint energetics are examined during terminal and non-terminal landing tasks to understand neuromuscular control strategies, and how this may influence injury risk potential (Decker et al., 2003, DeVita and Skelly, 1992, Schmitz and Shultz, 2010, Schmitz et al., 2002, Shultz et al., 2010 and Zhang et al., 2000). Landing is consistent with the type of movement often observed at the time of acute joint injury (Boden et al., 2000 and Olsen et al., 2004), and is a common laboratory research model used to understand the influence of potential risk factors associated with injury; for example, sex (Decker et al., 2003, Schmitz and Shultz, 2010 and Shultz et al., 2009), joint laxity (Shultz et al., 2010), and fatigue (Weinhandl et al., 2011), among others, on neuromechanical landing strategies. Using a landing research model, energetic analyses specifically examines the work done on/by the extensor muscles through integration of net joint powers (McNitt-Gray, 1993), which provides insight into the global biomechanical strategies that individuals may use to decelerate the body and dissipate impact forces (Zhang et al., 2000). However, individual laboratory methods can vary widely when using these landing models. In order to compare findings across studies, and the potential for these risk factors to influence injury potential during actual sport movement patterns, it is important to understand how the chosen research methods may influence the outcome measures of interest. Among these are the types of landing task employed and the shoe conditions chosen.

Landing energetics have been equally studied during the deceleration phase of terminal (drop landings = DL) (Decker et al., 2003, DeVita and Skelly, 1992, Schmitz et al., 2007 and Zhang et al., 2000) and non-terminal (drop jumps = DJ) tasks (Kovacs et al., 1999, McCaulley et al., 2007a, Moran and Wallace, 2007, Schmitz and Shultz, 2010 and Shultz et al., 2010) from multiple landing heights. While both tasks require the individual to reduce the body's momentum and absorb impact forces, the DJ requires the use of stored energy during the eccentric phase to maximize the mechanical work output of the subsequent concentric phase to reverse the body's direction. Because the concentric phase of movement is enhanced with greater eccentric work performed over a shorter range of motion (Moran and Wallace, 2007), energy dissipation strategies during the eccentric phase of a non-terminal DJ may be quite different from those of a terminal DL where force attenuation is the primary goal. This is suggested by one study (Shultz et al., 2010) that noted greater contributions at the hip and ankle relative to the knee during a 0.45 m DJ as compared to previous reports of unconstrained, simple DLs from similar heights (0.32–0.62 m) (Decker et al., 2003 and Zhang et al., 2000). Additionally, in a study examining muscle activation strategies between DJs and DLs, the landing phase of DJs were characterized by lower quadriceps and gastrocnemius activity prior to contact, and greater muscle activity following ground contact compared to drop landings (Ambegaonkar et al., 2011). These differences in task demands may have important research and clinical implications, as most sporting activities are non-terminal in nature, and studies of anterior cruciate ligament (ACL) injury indicate that injuries typically occur during an attempt to decelerate the body and change the direction of motion (Boden et al., 2000 and Krosshaug et al., 2007). Thus, while potential injury risk factors such as sex have routinely been examined for their effects on lower extremity biomechanics using both DL (Huston et al., 2001, Kernozek et al., 2005, Lephart et al., 2002 and Schmitz et al., 2007) and DJ (Dedrick et al., 2008, Ford et al., 2006, Shultz et al., 2009 and Shultz et al., 2010) landing tasks, results from non-terminal DJ tasks may ultimately be more revealing than terminal DL tasks when trying to understand one's ability to effectively dissipate forces during sport related activity. Thus, given the relatively equal use of DL and DJ in prior research to make these determinations, understanding how fundamental differences in these tasks influence the outcome measures of interest may aid our interpretation of findings across studies.

Another potential confound when interpreting landing energetics across studies is whether participants are shod or barefoot. Choice of shoe wear across studies ranges from utilizing standardized shoes (Decker et al., 2003) to the participant wearing their own shoe (DeVita and Skelly, 1992), to testing while barefoot (Schmitz and Shultz, 2010 and Shultz et al., 2010); some studies do not report shoe wear condition (Moran and Wallace, 2007 and Zhang et al., 2000). Barefoot testing has been used to control the potential confounding effect of different shoe structures, and has the added benefit of reducing movement artifact associated with placing sensors on shoes (Shultz et al., 2010). However, the extent to which barefoot landings cause participants to use a different landing strategy than when shod (the typical case during sport activity), is relatively unknown. While one study concluded that individuals may actually increase impact forces during a 4.5 cm step down when wearing shoes vs. barefoot in an attempt to improve foot-surface interface stability (Robbins and Waked, 1997), research comparing single leg landing biomechanics from a relatively low (15 cm) height with and without shoe wear reported no difference in landing mechanics in healthy limbs (Webster et al., 2004).

To date, we are not aware of any studies that have directly compared energy dissipation strategies between double leg DJ and DL when shod vs. barefoot. Such comparisons may yield a better understanding of system demands during terminal and non-terminal tasks, allowing more informed landing research designs and biomechanical interpretations in the future. For example, Decker et al. (2003) examined sex differences in joint energy absorption and associated kinematics and kinetics while subjects performed a terminal DL while wearing a standard court shoe while two companion studies by Schmitz and Shultz (2010) and Shultz et al. (2009) examined similar biomechanical variables while subjects performed a non-terminal DJ while barefoot. Energetic findings between the studies differed at the hip and ankle, as did the associated kinematics and kinetics (initial and peak joint excursions, peak internal joint moments) of the observed energetics. While there are potentially many reasons for the differential findings, yet unknown is whether the observed differences are simply due to methodological differences, thus making comparisons across various landing studies more challenging.

Therefore, we compared sagittal plane energy absorption and torsional stiffness/impedance during the initial landing phase of DJs and DLs, and between shod and barefoot conditions, in order to better understand how methodological factors alter landing strategies (and therefore subsequent interpretation of study findings). To further interpret differences in energy absorption and torsional stiffness/impedance between task and shoe conditions, associated kinematics (initial contact and total excursion joint angles), kinetics (ground reaction force, peak joint moments) and muscle activation (pre- and post-landing) amplitudes were also examined. For our primary outcome variables, we expected greater energy absorption during the DJ compared to the DL, and lower torsional joint stiffness/impedance when barefoot as compared to the shod conditions.

2. METHODS

Twenty (10 F, 10 M) college students (21.3 ± 2.9 yrs, 173.7 ± 7.0 cm, 71.2 ± 9.2 kg) with no lower extremity injury in the past 6 months, and who were active in jumping and landing activities 3×/week for at least one sport season participated. All participants signed a consent form approved by the University's institutional review board, and were familiarized to all testing procedures 2 days prior to testing. All analyses were completed on the dominant limb, defined as the preferred stance limb when kicking a ball.

After preparing the skin, 10 mm bipolar Ag–AgCl surface electrodes (Blue Sensor N-00-S; Ambu Products, Ostykke, Denmark) were placed in parallel (center to center distance 20 mm) between the motor point and the distal tendon of the vastus lateralis, vastus medialis, biceps femoris, semimembranosus, and medial and lateral gastrocnemius heads, and oriented longitudinal to the muscle fibers per normal anatomy (Basmajian 1980). For normalization purposes, surface electromyography (sEMG) was recorded during 3, 5-s maximal effort isometric contractions (MVICs) of the quadriceps, hamstrings and gastrocnemius muscles using a 16 channel Myopac telemetric system (Run Technologies, Mission Viejo, CA; internal hardware sampling rate 8 kHz, amplification 1 V/mV, frequency bandwidth 10–1000 Hz, common mode rejection ratio 90 dB min at 60 Hz, input resistance 1 MΩ). Quadriceps and hamstring MVICs were performed at 25° knee flexion against an instrumented dynamometer (Biodex Medical Systems Inc., Shirley, NY). Gastrocnemius MVICs were performed at 25° knee flexion and 10°ankle plantar flexion against a non-elastic strap.

Electromagnetic sensors (Ascension Technologies, Burlington, VT) were then attached over the anterior mid-shaft of the third metatarsal, the mid-shaft of the medial tibia, the mid-shaft of the lateral femur, the sacrum and the C7 spinous process. Joint centers were determined using Leardini et al. (1999) (hip) and centroid (Madigan and Pidcoe 2003) (knee and ankle) methods. Kinematic (100 Hz), kinetic (1000 Hz) and sEMG (1000 Hz) data were simultaneously acquired (10 N foot contact threshold for trigger onset) during 10 successful DJs (five barefoot/five shod) and DLs (five barefoot/five shod) from a 0.45 m platform positioned 0.1 m behind the rear edge of the force plate (Type 4060-nonconducting; Bertec Corporation, Columbus, OH). With the hands positioned at ear level to control for the influence of upper limb motion on task outcomes (Laffaye et al., 2006), and the toes aligned over the leading edge of the platform, subjects were instructed to fall off of the box, without jumping or stepping, and either (a) land with both feet simultaneously and return to a standing position (DL) (Riemann et al., 2002) or (b) land with both feet simultaneously and immediately performing a maximum double leg vertical jump (DJ) (Shultz et al., 2009). The order of both landing tasks and shoe conditions were equally counterbalanced using a Latin square design. No instructions were provided on how to land to ensure subjects used their preferred landing style and to prevent experimenter bias. For the shod condition, subjects wore their own self-selected running shoe. Given the within-subject study design, we chose to not standardize the make and model of shoe to allow greater generalization of our findings. Participants were given sufficient practice to become comfortable with each landing condition prior to testing. Trials were discarded and repeated if the subject stepped or jumped off the box, contacted the ground with asynchronous foot contact, or landed with the test leg off of the intended force plate.

2.1. Data reduction and analyses

SEMG signals were band pass filtered from 10 to 350 Hz using a fourth order, zero-lag Butterworth filter, rectified, then processed using a root mean square algorithm with a 100 ms (MVIC trials) or 25 ms (landing trials) time constant. The peak (root mean square) RMS amplitudes obtained for each muscle during the 150 ms immediately before (pre) and the 150 ms after (post) initial ground contact were averaged across the five trials for each condition, and normalized to their respective peak MVIC trials (%MVIC) (Shultz et al., 2009). Values from the medial and lateral aspect of each muscle group were averaged and expressed as a single %MVIC.

Kinematic data were linearly interpolated to the sampled kinetic data and low pass filtered at 12 Hz using a fourth order, zero-lag Butterworth filter (Bisseling and Hof, 2006). Peak vertical ground reaction forces (vGRF) were normalized to body weight (%BW). A segmental reference system was defined for all body segments. Positive z-axis was defined the left-to-right axis (right direction positive), the positive y-axis was defined as the distal-to-proximal longitudinal axis (proximal direction positive), and the positive x-axis defined as the posterior-to-anterior axis (anterior direction positive). Hip, knee, and ankle flexion angles were calculated using Euler angle definitions with a rotational sequence of $Z\ Y'\ X''$ and hip, knee, and ankle intersegmental moments were calculated via inverse dynamics (Motion Monitor Software; InnSport, Chicago, IL) and normalized to each subject's weight and height ($N\ m \times BW^{-1} \times Ht^{-1}$). Sagittal plane hip, knee, and ankle torsional joint stiffness was calculated for DJ (Schmitz and Shultz, 2010) while sagittal plane hip, knee, and ankle torsional joint impedance was calculated for DL (Kulas et al., 2006). While both stiffness and impedance were calculated as change in normalized net internal moment divided by change in angular position from initial contact to peak flexion excursion, normalized to %body weight (N) and height (m) (N m × BW^{-1} × Ht^{-1} × deg^{-1}), we differentiate this terminology as stiffness is mechanically associated with energy-conservative tasks such as DJ whereas impedance is associated with energy-dissipative capacity tasks such as DL (Kulas et al., 2006). Work done on the extensor muscles (representing energy absorption) was then calculated by integrating the negative portion of the joint power curve (J × BW^{-1} × Ht^{-1}), with joint power defined as the product of the normalized joint moment and joint angular velocity at each time point. For reporting purposes, all data were average across the five trials for each condition, and flexion defined as the positive direction.

A 2 (DJ, DL) × 2 (shod, barefoot) repeated measures ANOVA examined differences in vertical ground reaction forces. A multivariate 2 (DJ, DL) × 2 (shod, barefoot) × 3 (hip, knee, ankle) repeated measures ANOVA examined differences in joint energy absorption, stiffness, peak moments, initial joint angles, and joint excursions (peak-initial). A separate multivariate 2 (DJ, DL) × 2 (shod, barefoot) × 3 (quadriceps, hamstring, gastrocnemius) repeated measures ANOVA examined differences in pre- and post-landing muscle activation amplitudes. Significant multivariate results (conducted initially to control for Type I error) were then followed by univariate tests and post hoc pair-wise comparisons (Bonferroni corrected) to identify significant differences within each variable. Alpha level was set at $P \leqslant 0.05$. All analyses were performed using PASW Statistics (V. 18.0.2; IBM Corp., New York). Conservatively estimating a correlation of 0.5 among biomechanical variables in the multivariate model with 20 subjects, we had 80% power to detect a moderate effect (0.30) with an alpha level of $P \leqslant .05$ [G ∗ Power 3 (Faul et al., 2007)].

3. RESULTS

Peak vGRF was larger during DL vs. DJ (P = .001), and during shod vs. barefoot conditions (P = .005), with the magnitude of difference being greater for DL (1.92 ± 0.40 vs. 1.75 ± 0.26 BW; P = .005) compared to DJ (1.67 ± 0.27 vs. 1.61 ± 0.27 BW; P = .053) (P = .028). Table 1 reports values for energy absorption, torsional stiffness/impedance, peak extensor moments, and initial and excursion angles stratified by joint, task and shod condition. Multivariate statistics were significant for all main effects [task (P = .014), shod (P = .044), joint (P < .001)] and interactions [task by shod (P = .034), task by joint (P = .016), shod by joint (P < .001) and task by shod by joint (P = .014)]. Univariate results are summarized as follows.

Table 1
Biomechanical variables (Mean ± SD) stratified by joint, task and shoe condition.

	Drop jump		Drop landing	
	Shod	Barefoot	Shod	Barefoot
Energy absorption ($J \times BW^{-1} \times Ht^{-1}$)[a]				
Hip[b]	−7.99 ± 3.20	−8.48 ± 3.30	−6.02 ± 3.10	−7.03 ± 2.80
Knee	−3.21 ± 1.60	−3.22 ± 1.70	−3.16 ± 1.60	−3.33 ± 1.70
Ankle	−7.86 ± 2.30	−8.76 ± 3.30	−8.23 ± 2.20	−9.06 ± 4.00
Joint torsional stiffness/impedance ($N\,m \times BW^{-1} \times Ht^{-1} \times deg^{-1}$)[a]				
Hip	−0.59 ± 0.35	−0.50 ± 0.26	−0.89 ± 0.77	−0.50 ± 0.21[c]
Knee	−0.14 ± 0.06	−0.13 ± 0.07	−0.14 ± 0.06	−0.14 ± 0.06
Ankle	−0.17 ± 0.04	−0.21 ± 0.05[c]	−0.19 ± 0.06	−0.21 ± 0.08
Peak extensor moments ($N\,m \times BW^{-1} \times Ht^{-1}$)				
Hip	−0.23 ± 0.04	−0.22 ± 0.03	−0.24 ± 0.05	−0.21 ± 0.04[c]
Knee	−0.07 ± 0.03	−0.07 ± 0.03	−0.07 ± 0.02	−0.07 ± 0.03
Ankle	−0.11 ± 0.02	−0.12 ± 0.03[c]	−0.12 ± 0.03	−0.12 ± 0.03
Initial flexion/plantar flexion angle (deg)				
Hip[b]	17.6 ± 9.3	18.0 ± 5.9	13.3 ± 8.1	12.7 ± 6.9
Knee[b]	13.4 ± 6.5	12.8 ± 5.8	7.4 ± 8.5	9.0 ± 5.6
Ankle	40.3 ± 4.9	42.8 ± 6.6	39.1 ± 6.7	41.1 ± 6.2
Total joint flexion excursion (deg)				
Hip[b]	41.5 ± 13.9	41.8 ± 13.1	34.9 ± 16.8	39.0 ± 14.7
Knee[d]	69.9 ± 10.7	70.4 ± 11.7	66.5 ± 14.9	70.4 ± 13.0
Ankle[d]	62.4 ± 7.3	58.1 ± 9.2	60.9 ± 8.6	58.5 ± 7.9
Ground reaction force (BW)[b]	1.67 ± 0.27[c]	1.61 ± 0.27	1.92 ± 0.40[c]	1.75 ± 0.26[e]

[a] Values expressed as $\times 10^2$.
[b] DJ ≠ DL (regardless of footwear condition).
[c] Shod ≠ barefoot within task.
[d] Shod ≠ barefoot (regardless of task).
[e] Difference in shod vs. barefoot greater in DL vs. DJ.

3.1. Energy absorption

Significant results for task ($P = .013$), joint ($P < .001$) and task by joint interaction ($P = .002$) revealed that energy absorption was greater in the hip, but similar for the knee and ankle during DJ compared to DL (Fig. 1). This resulted in the hip, knee and ankle contributing 41.6%, 16.2% and 42.0%, respectively, to the DJ, and 35.4%, 17.6% and 47.0%, to the DL. Shoe condition had no effect on energy absorption (*P*-values for main effect and interactions = .086 to .783).

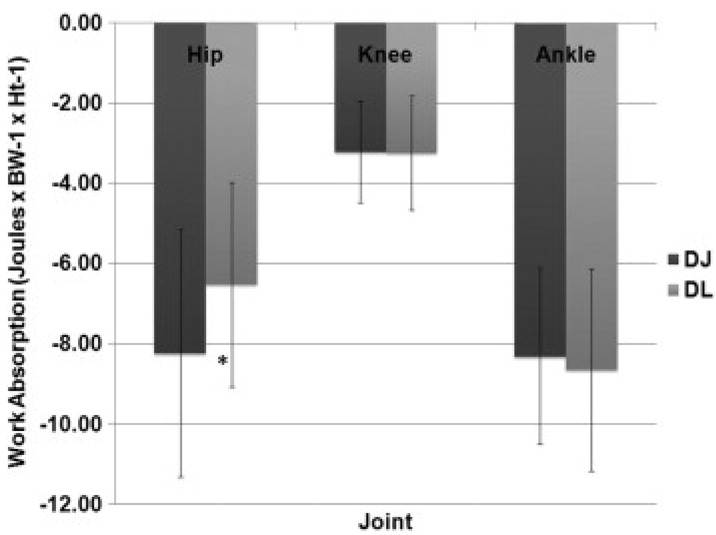

Fig. 1.
Energy absorption at the hip, knee and ankle during drop jumps (DJ) and drop landings (DL) (*DJ ≠ DL).

3.2. Joint stiffness/impedance

Stiffness/impedance (N m × BW^{-1} × Ht^{-1} × deg^{-1}) was generally greater during shod (−0.35 ± 0.03) compared to barefoot conditions (−0.28 ± 0.08) ($P = .036$); however, this difference varied by task ($P = .024$), joint ($P = .001$) and task by joint (3-way interaction, $P = .011$, Table 1). Pairwise comparisons revealed greater hip impedance/stiffness for the shod compared to barefoot condition in the DL ($P = .018$) but not DJ ($P = .063$), while ankle stiffness/impedance was greater in the barefoot compared to shod condition for DJ ($P = .012$) but not DL ($P = .337$). Knee stiffness/impedance was unaffected by shoe and task ($P = .911$).

3.3. Peak extensor moments

Similar to joint stiffness/impedance, peak extensor moments varied between shoe conditions by task ($P = .009$), joint ($P = .016$), and task by joint (3-way interaction, $P = .010$, Table 1). Peak hip extensor moments were greater when shod vs. barefoot for DL ($P = .001$) but not DJ ($P = .005$), while peak ankle extensor moments were greater when barefoot vs. shod for the DJ ($P = .045$) but not DL ($P = .479$). This resulted in greater peak ankle extensor moments for the DL vs. DJ when shod ($P = .047$).

3.4. Initial joint angles

Joint angles at initial ground contact were more flexed for DJ (24.2 ± 4.2°) compared to DL (20.4 ± 4.6°) (task main effect; $P < .001$). Initial joint angles were smaller for the hip and knee in the DL compared to DJ (task by joint interaction, $P = .012$, Fig. 2).

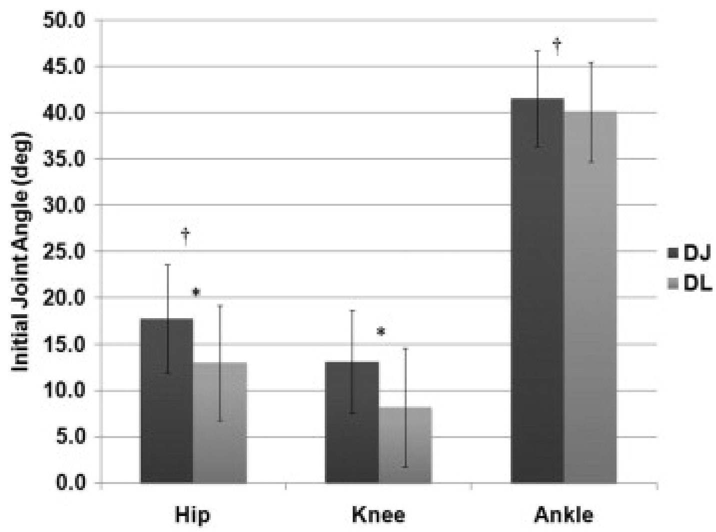

Fig. 2.
Initial joint angles by task and joint (*DJ ≠ DL, †ankle > hip > knee).

3.5. Joint excursions

Hip joint excursion was greater during DJ (41.7 ± 13.0°) compared to the DL (36.9 ± 14.9°), while knee (70.1 ± 11.0° vs. 68.5 ± 13.6°) and ankle (60.2 ± 8.0° vs. 59.7 ± 7.8°) excursions were similar between tasks (task by joint interaction, P = .049). Conversely, when comparing shoe conditions, there was no difference in hip joint excursion (38.2 ± 14.6° vs. 40.4 ± 13.0°), but there was less knee excursion (68.2 ± 12.1° vs. 70.4 ± 11.3°) and greater ankle excursion (61.6 ± 7.6° vs. 58.3 ± 8.4°) in the shod condition (shoe by joint interaction, P = .001). Consequently, total joint excursions were greater during DJ vs. DL when shod (57.9 ± 9.0° vs. 54.1 ± 12.1°) but not when barefoot (56.8 ± 9.6°vs. 55.9 ± 10.4°) (task by shoe interaction; P = .031).

3.6. Pre- and post-landing muscle activation amplitudes

Table 2 reports pre- and post-landing muscle activation data stratified by muscle, task and shod condition. Multivariate results revealed differences in muscle activation by task (P = .001), shod (P = .008) and muscle (P < .001), and the interactions of task by muscle (P = .002) and shoe by muscle (P = .014), but not for task by shoe (P = .637) or task by shoe by muscle (P = .413). Post hoc analyses indicated that in preparation for landing, both quadriceps (14.9% vs. 11.6%) and gastrocnemius (38.3% vs. 35.8%) muscle activation amplitudes were greater during the DL than DJ (P = .005 and .028, respectively), with no difference observed in the hamstrings (11.4 vs. 11.0, P = .433). Upon ground contact, quadriceps (68.8% vs. 84.0%), hamstring (22.1 vs. 24.5) and gastrocnemius (19.0% vs. 28.1%) activations were all lower during the DL than DJ (P < .037) (Fig. 3). When examining the effects of shoe wear, there was greater hamstring (12.1 ± 3.8 vs. 10.3 ± 4.2%, P = .006), but similar quadriceps (13.3 ± 4.3% vs. 13.2 ± 5.0%, P = .995) and gastrocnemius (36.2 ± 13.1% vs. 37.9 ± 16.3%, P = .156) muscle activation in preparation for ground contact, and less overall muscle activation after ground contact (39.2% vs. 43.0%, P = .002) in barefoot vs. shod conditions.

Table 2
Mean ± SD for pre- and post-landing muscle activation amplitudes (%MVIC) stratified by muscle, task and shoe conditions.

	Drop jump		Drop landing	
	Shod Mean ± SD	Barefoot Mean ± SD	Shod Mean ± SD	Barefoot Mean ± SD
Pre-landing				
Quadriceps[a]	11.1 ± 3.1	12.1 ± 3.8	15.4 ± 7.9	14.4 ± 5.1
Hamstring[b]	10.1 ± 4.7	11.9 ± 3.5	10.5 ± 3.9	12.3 ± 4.4
Gatrocnemius[a]	36.7 ± 16.1	34.8 ± 12.8	39.0 ± 17.1	37.6 ± 14.1
Post-landing[b]				
Quadriceps[a]	87.4 ± 37.7	80.9 ± 33.9	70.9 ± 29.4	66.8 ± 27.1
Hamstring	26.4 ± 14.2	22.6 ± 13.4	24.1 ± 15.2	20.7 ± 13.2
Gatrocnemius[a]	29.6 ± 14.0	26.7 ± 11.7	19.8 ± 6.5	18.3 ± 6.9

[a] DJ ≠ DL (regardless of footwear condition).
[b] Shod ≠ barefoot (regardless of task).

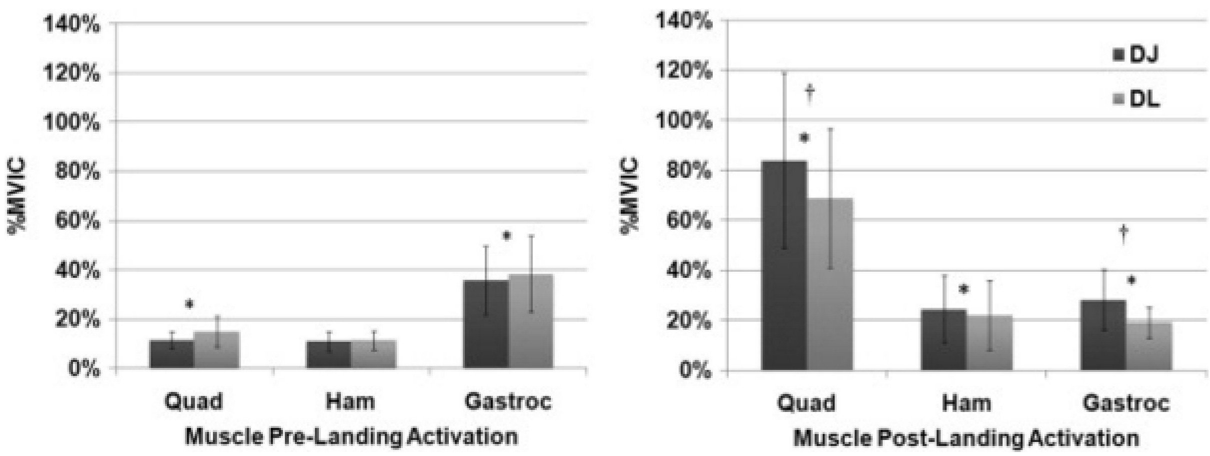

Fig. 3.

Pre- (a) and post (b) landing muscle activation during drop jumps (DJ) and drop landings (DL) (*DJ ≠ DL, †Difference between DJ and DL greater than hamstrings).

4. DISCUSSION

Our primary findings were that energy absorption was greater in the DJ than the DL, primarily due to greater energy absorption at the hip (modulated by an increase in hip joint excursion) and this was irrespective of shoe condition. Conversely, torsional joint stiffness/impedance was primarily affected by shoe condition, with hip impedance/stiffness substantially greater when shod vs. barefoot during the DL but not during DJ, and with ankle stiffness/impedance greater when barefoot vs. shod during DJ but not during DL.

4.1. Comparison of task differences

Understanding fundamental differences in terminal and non-terminal landing tasks may further inform our choice of laboratory tasks and the subsequent interpretation of research findings. The fundamental difference between DJ and DL is that subjects landed in a more flexed position and absorb more energy (primarily through the hip extensors) in the DJ, a strategy associated with greater mechanical efficiency (McCaulley et al., 2007b) and subsequent jump height (Moran and Wallace, 2007). This greater energy absorption in DJ vs. DL was accompanied by greater initial flexion angles in hip and knee, and greater overall muscle activation upon ground contact. This may represent a strategy to prepare for ensuing activity by maximizing the force–length relationship of the extensor musculature (Visser et al., 1990) and optimizing elastic properties of tendinous structures (Kubo et al., 2007). Conversely, the factors of greater activation of the quadriceps and gastrocnemius muscles in preparation for landing, more extended joints, less hip absorption, and greater vGRF are indicative of a “stiffer” landing style to decelerate the body’s momentum during DL where greater impact forces are potentially absorbed by other joints, or by passive structures. While we are aware of no other studies directly comparing energetics between

DJ and DL, the greater quadriceps and gastrocnemius pre-landing activation, greater vGRF and lower post-landing activation in the DL vs. DJ is consistent with a previous study comparing muscle activation strategies (Ambegaonkar et al., 2011).

Moreover, there appears to be a shift from a more proximal energy absorption strategy in DJ to a more distal strategy in DL. While relative energy absorption was not a primary variable of the current study, qualitative comparisons revealed greater relative hip absorption in the DJ (41.6% vs. 35.4%) and greater ankle relative absorption in the DL (46.9% vs. 42.1%), with little difference in relative knee energy absorption between tasks (16.2% vs. 17.6%). The greater relative demands at the hip in the DJ vs. DL appear to be in line with prior findings during unconstrained 0.45 m DJ (Shultz et al., 2010) during which the hip contributed 40% of total energy absorption as compared to only 30–34% of the total energy absorption during DLs from similar heights (0.32–0.62 m) (Decker et al., 2003 and Zhang et al., 2000). However, our findings of greater relative contributions at the ankle appear to be inconsistent with prior research as the ankle contributed 20–36% to total energy absorption during DLs (Decker et al., 2003 and Zhang et al., 2000) but contributed as much as 45% in another DJ study (Shultz et al., 2010). This may be due to the fact that current participants may have landed in a more “stiff” manner as suggested by the greater relative contribution of the ankle to energy absorption (Zhang et al., 2000). Other potentially confounding factors include the balance of males and females in the respective studies (Decker et al., 2003 and Zhang et al., 2000) as well as differences in landing height (0.45 m in current study but 0.32 and 0.62 m (Zhang et al., 2000) and 0.60 m (Decker et al., 2003) in others).

Because the majority of sporting activities are non-terminal in nature, and because more acute joint injuries occur as an individual performs one task while readying themselves for another (e.g. attempting to change direction of motion) (Boden et al., 2000 and Olsen et al., 2004), the substantial differences in energy absorption strategies between DJ and DL suggest that the DJ may provide a more realistic laboratory model when examining the global biomechanical strategies used by individuals to decelerate the body and dissipate impact forces during landing activities commonly associated with injury. Although the typical landing biomechanics of the DJ would generally considered being “protective” relative to typical drop landing tasks, injuries still occur during these non-terminal tasks during sporting activity. Hence, using a task that better mimics sport demands at the time of injury may help us better elucidate how relevant factors (e.g. gender, joint laxity, and fatigue) influence biomechanics during these motions. Moreover, given recent interest in the role of proximal hip control on distal lower extremity motion (McLean et al., 2005, Pollard et al., 2007 and Sigward and Powers, 2007), using tasks that place more energy demands on the hip musculature may better expose these potential relationships.

4.2. Effect of shod condition

Although much attention has been given to the role of shoe wear in attenuating impact forces during running (Bishop et al., 2006 and Lieberman et al., 2010), less is known about its effect on landing activities (Webster et al., 2004 and Yeow et al., 2011). Our results indicate that shoe wear has little impact on energy absorption strategies during DJ vs. DL tasks, but has considerable influence on torsional stiffness/impedance. Lower overall stiffness/impedance in the barefoot vs. the shod condition is consistent with the lower post-landing muscle activation amplitudes in the barefoot vs. shod conditions. Further, this lower stiffness/impedance during barefoot landings was primarily driven by the substantially lower hip impedance when barefoot during the DL, even though ankle stiffness was somewhat greater when barefoot during the DJ. Our results are counter to a recent study investigating the effects of shoe wear on knee biomechanics during DL (Yeow et al., 2011). While they reported increased knee flexion excursion, moments, and energy absorption when shod vs. barefoot our study revealed no difference in peak knee extensor moments or energy absorption, and smaller knee flexion excursions when shod during the DL. The use of standardized shoes in the previous investigation as compared to the self-selected shoes in the current study may be one potential source for the different findings. That is, if participants in the previous study were not completely accustomed to the standardized footwear, this could have potentially affected their movement patterns. However, it is difficult to fully compare findings between studies as the previous investigation was limited to the reporting of knee joint biomechanics.

Conversely, our findings closely mirror those of the running literature. A study comparing habitual barefoot and shod runners revealed that barefoot runners adopted a strategy of forefoot striking (more plantarflexed ankle) while shod runners adopted a rearfoot striking strategy (Lieberman et al., 2010). Although the demands of running and landing differ, control of center of mass positioning is needed in landing (McNitt-Gray et al., 2001) as well as running (Moritz and Farley, 2006). Center of mass vertical position during functional tasks is largely influenced by lower extremity flexion angle (Farley and Morgenroth, 1999). As the current study revealed an increase in knee flexion excursion and decrease in ankle dorsiflexion excursion, and lower overall vertical ground reaction forces and lower extremity muscle activation upon ground contact when barefoot vs. shod, a multi-joint strategy is likely implemented to handle impact forces depending on conditions. Specifically, the decreased dorsiflexion excursion when barefoot may be a distal strategy to maintain increased ankle stiffness in preparation for the subsequent propulsive phase while also reducing overall impact forces through lower overall muscle activation and increased excursion at the knee. Collectively, these data suggest that shoe wear may influence the mechanics of the lower extremity during landing from a jump as the body adopts different mechanical strategies to both control peak impact forces and prepare for subsequent activity, without adversely impacting joint specific energy absorption strategies. This emphasizes the importance of carefully considering the shoe condition when examining stiffness/impedance in future studies, which may differ depending on the population of interest and whether they are habitually barefoot (e.g. dancers) vs. shod (e.g. basketball players) during activity.

One potential limitation of this study is that we did not control the type of running shoe each subject wore. Because this was purely a within-subject study design, we chose to let participants wear shoes to which they were already habituated so as not to influence their natural landing strategies. While this may have potentially contributed to the greater variability observed in hip stiffness/impedance and initial joint angles compared to barefoot conditions, this greater variability was not observed across all measurements or joints and did not prevent us from identifying meaningful differences between shoe conditions, including the hip variables noted above. Although we believe our results based on self-selected shoes allows greater generalization of our findings, future studies should consider the extent to which variations in shoe type contribute to individual variations in energy absorption strategies, particular given the difference in our findings compared to others using a standardized shoe (Yeow et al., 2011). Another potential limitation is that we did not instrument both limbs, thus our verification of balanced, bilateral landing trials was limited to our visual observations of synchronous foot contacts. While the potential for asynchronous landings may have introduced some random variability in our data, we have no reason to suspect this would have affected one condition more than another.

5. CONCLUSIONS

To our knowledge the current study is the first investigation to determine energetic differences in terminal vs. non-terminal landing tasks, and the influence of acute shoe wear condition on hip, knee, and ankle energetics and stiffness/impedance associated with each task. Because the majority of sporting activities are non-terminal in nature, the substantial differences observed in energy absorption strategies between DJ and DL suggest that the DJ may provide a more realistic laboratory model to examine how relevant factors (e.g. sex, joint laxity, fatigue) influence global biomechanical strategies when attempting to decelerate the body and dissipate impact forces during activities commonly associated with injury. While we observed no effect of shoe wear on energy absorption strategies during DL or DJ (suggesting that comparison of studies across different shoe wear conditions may be valid), shoe wear should be carefully considered when examining torsional joint stiffness/impedance, as these outcomes were acutely modulated in response to the amount of cushioning available.

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