The influences of sex and posture on joint energetics during drop landings

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Abstract:

Previous observations suggest that females utilize a more erect initial landing posture than males with sex differences in landing posture possibly related to sex-specific energy absorption (EA) strategies. However, sex-specific EA strategies have only been observed when accompanied by sex differences in initial landing posture. This study (a) investigated the potential existence of sex-specific EA strategies; and (b) determined the influences of sex and initial landing posture on the biomechanical determinants of EA. The landing biomechanics of 80 subjects were recorded during drop landings in Preferred, Flexed, and Erect conditions. No sex differences in joint EA were identified after controlling for initial landing posture. Males and females exhibited greater ankle EA during Erect vs Flexed landings with this increase driven by 12% greater ankle velocity, but no change in ankle extensor moment. No differences in hip and knee EA were observed between conditions. However, to achieve similar knee EA, subjects used 7% greater mean knee extensor moment but 9% less knee angular velocity during Flexed landings. The results suggest that sex-specific EA strategies do not exist, and that the magnitude of knee joint EA can be maintained by modulating the relative contributions of joint moment and angular velocity to EA.

Keywords: Kinematics | kinetics | energy absorption | landing biomechanics

Article:

Compared with males, females are at significantly greater risk for patellofemoral pain syndrome (PFPS; Taunton et al., 2002; Boling et al., 2010) and anterior cruciate ligament (ACL) injury (Arendt et al., 1999; de Loës et al., 2000; Agel et al., 2005). Females also tend to exhibit a more erect posture during landing (Malinzak et al., 2001; Decker et al., 2003; Yu et al., 2006) with lesser knee flexion at ground contact serving as an identified risk factor for the development of PFPS (Boling et al., 2009) and a potential contributor to increased ACL loading (Nunley et al., 2003). Consequently, it has been suggested that PFPS and ACL injury prevention programs include components specifically targeted at increasing knee flexion during landing (Hewett et al., 1999; Myklebust et al., 2003; Mandelbaum et al., 2005; Boling et al., 2009). However, despite these recommendations, the biomechanical reason(s) underpinning the use of a more erect landing position by females remain unknown.

Decker et al. (2003) postulated that sex differences in landing posture may be driven by sex-specific sagittal plane energy absorption (EA) strategies in which males and females preferentially use either the hip or ankle, respectively, in conjunction with the knee as the primary joints with which to absorb energy during landing. They proposed that the more erect landing posture of females in their investigation was the result of a female-specific ankle and knee joint dominant EA strategy, and that the use of an erect landing posture with this strategy served to maximize the amount of energy females absorbed at these joints during landing (Decker et al., 2003). However, it remains unclear whether these proposed sex-specific EA strategies truly exist as sex differences in joint EA magnitudes and joint contributions to EA (e.g., EA strategy) have only been reported when accompanied by sex differences in initial landing posture (Decker et al., 2003; Norcross et al., 2010a).

The magnitude of joint EA can be affected by changes to either of the two biomechanical determinants of joint EA – angular velocity or resultant joint moment (Winter, 2005). Accordingly, Mizrahi and Susak (1982) proposed that increasing the total available joint ranges of motion during landing by positioning the joints in lesser flexion at initial ground contact might increase the ability of muscles spanning these joints to absorb energy by potentially allowing for greater joint angular velocities. This notion is supported by Zhang et al. (2000) who observed greater magnitudes of EA during landings with greater hip, knee, and ankle angular displacements. It is plausible, therefore, that the magnitude and distribution of energy absorbed by individual joints during landing (i.e., EA strategies) is influenced by initial contact joint configurations, rather than feedforward EA strategies dictating the use of particular initial contact joint configurations. As such, observed sex differences in landing posture may be driven by other sex-related biomechanical factors, such as strength, and not sex-specific EA strategies (Lephart et al., 2002). For instance, females might adopt a more erect landing posture in order to achieve adequate joint EA through the utilization of greater joint angular velocities, but lesser net joint moments (e.g., joint extensor moment demands).

Given these possibilities, the objective of this study was to further investigate the potential existence of sex-specific EA strategies that could contribute to sex differences in landing posture by evaluating the influences of sex and landing posture on joint EA. We also sought to examine the biomechanical determinants of EA to elucidate whether the mechanisms through which EA is achieved (i.e., joint angular velocity and moment) are influenced by sex and landing posture. We hypothesized that compared with males, females would demonstrate a more erect landing posture and an ankle/knee dominant EA strategy when completing drop landings using a preferred posture, but that no sex differences in joint EA would be identified after experimentally controlling for landing posture (i.e., during constrained flexed and erect landing conditions). Further, we hypothesized that males and females would utilize greater joint angular velocity, but lesser joint moment to achieve similar magnitudes of individual joint EA when landing in an erect posture compared with a flexed posture.

METHODS

Subjects

Eighty physically active (40 women and 40 men) volunteers were recruited for participation in this study. All subjects were recreationally active (participating in at least 30 min of physical activity at least three times per week), and generally healthy with no history of ACL injury, neurological disorder, lower extremity surgery, or lower extremity injury within the 6 months prior to data collection. The investigation was approved by the University's Institutional Review Board and all subjects provided written informed consent prior to participation.

Subject preparation

Prior to data collection, the height and mass of each subject were recorded and used for generation of the biomechanical model and normalization of the dependent variables. An electromagnetic motion capture system (MotionStar, Ascension Technology Corp., Burlington, Vermont, USA) and five 6 degree of freedom electromagnetic tracking sensors were used to assess dominant lower extremity and trunk kinematics. Sensors were positioned over the third metatarsal, anteromedial shank, and lateral thigh of the dominant leg (defined as the leg used to kick a ball for maximum distance), as well as the sacrum and C7 spinous process. Sensors were placed over areas of minimal muscle mass and secured with prewrap and athletic tape in order to reduce motion artifact. Global and segmental axis systems were established using a right-hand coordinate system with the positive X-axis designated as forward/anteriorly, the positive Y-axis leftward/medially, and the positive Z-axis upward/superiorly. The MotionMonitor motion analysis software (Innovative Sports Training, Inc., Chicago, Illinois, USA) was used to create a link-segment model of the dominant lower extremity, pelvis, and thorax by digitizing the ankle, knee, and hip joint centers and the T12 spinous process. Ankle and knee joint centers were defined as the midpoints of the digitized medial and lateral malleoli, and the medial and lateral femoral condyles, respectively. The hip joint center was predicted using external landmarks on the pelvis (Bell et al., 1989). Finally, a nonconductive force plate (Type 4060-NC, Bertec Corporation, Columbus, Ohio, USA), whose axis system was aligned with the global axis system, was used to measure reaction forces and moments during the drop landing trials.

Preferred drop landings

Following setup, subjects completed double-leg terminal drop landings from a height of 0.60 m in three different landing postures: Preferred, Flexed, and Erect. All subjects completed the Preferred condition first to eliminate the possibility that their preferred landing strategy would be contaminated by completing the constrained Flexed and Erect landing conditions. For the Preferred condition, drop landings were initiated from atop a 0.60 m tall box positioned directly behind the force plate in order to precisely replicate the task used by Decker et al. (2003). Subjects were instructed to reach out with their dominant foot to position it over the force plate; roll forward off of the box using their nondominant foot without jumping or lowering themselves in order to initiate a drop; and then to perform a double-leg terminal landing with their dominant foot positioned completely on the force plate and their nondominant foot positioned on the floor next to the force plate. Subjects were given no other instructions or feedback regarding landing technique or performance. All subjects performed three practice trials followed by five testing trials in the Preferred condition before completing drop landings in the constrained conditions.

Flexed and Erect drop landings

In order to experimentally manipulate knee flexion angle at initial contact during drop landings, it was necessary to have subjects hang from an overhead bar attached to a wooden support frame (Fig. 1) and provide them with real-time biofeedback regarding their knee flexion angle using the MotionMonitor system and a computer monitor. Biofeedback was presented on the monitor in the form of a cursor and target window that helped subjects to position their knees in 35 ± 5° and 20 ± 5° of flexion, respectively, during the Flexed and Erect landing conditions. These target knee angles were chosen as they are similar to the mean knee flexion angles at initial contact exhibited by male (Flexed) and female (Erect) subjects in a previous study that demonstrated sex differences in EA strategy during 0.60 m drop landings (Decker et al., 2003). Though subjects only received feedback regarding the knee flexion angle of their dominant leg, they were instructed to move both legs in unison. Once subjects successfully positioned the cursor within the target window to achieve the desired knee flexion angle, an auditory signal was triggered. Subjects were instructed that they could let go of the bar to initiate the drop whenever they were ready so long as the auditory signal was present. They were then to maintain their body position until the instant of ground contact at which time they could move their joints in whatever manner they chose in order to complete the double-leg drop landing.

Figure 1. Experimental setup and drop bar used during the constrained landing conditions.

Impact velocity was standardized during the constrained conditions by adjusting the overhead bar to maintain the ankle joint center at approximately 0.60 m. The bar was initially positioned using an algorithm developed during pilot testing based upon subject height and the expected hip joint angles that subjects would need to use to position their feet under their center of mass while flexing the knee to the angles desired in each constrained landing condition. After the initial adjustment, drop height was verified by monitoring the vertical position of each subject's ankle joint center just prior to the initiation of the drop during each condition's practice trials. If necessary, the drop bar was further adjusted to achieve consistent 0.60 m drop landings across all conditions.

After each trial in the Flexed and Erect conditions, knee flexion angle and vertical ground reaction force were immediately calculated and displayed. These data were used to determine the knee flexion angle at the instant of ground contact and trials were deemed successful if this value was within the prescribed ranges for each experimental condition. All subjects completed a minimum of three practice trials, but were restricted to attempting a maximum of eight Flexed and Erect testing trials in hopes of capturing five successful trials for each condition. Subjects were provided with at least 30 s of rest between trials and 2 min of rest between conditions to minimize the potential effects of fatigue, and the order of Flexed and Erect landings was counterbalanced across subjects.

Data sampling and reduction

Kinematic and kinetic data were sampled at 120 Hz and 1200 Hz, respectively, using the MotionMonitor software. Raw kinematic data were low-pass filtered using a fourth-order, zero-phase-lag Butterworth filter with a cutoff frequency of 10 Hz, time synchronized with the kinetic data, and then resampled to 1200 Hz. Joint angular positions were calculated based on a right-hand convention using Euler angles in a Y (flexion/extension), X' (adduction/abduction), Z'' (internal/external rotation) rotation sequence with motion defined about the hip as the thigh relative to the pelvis, about the knee as the shank relative to the thigh, and about the ankle as the foot relative to the shank. Instantaneous joint angular velocities were calculated as the first derivative of angular position. Kinetic data were also low-pass filtered at 10 Hz (fourth-order zero-phase lag Butterworth) and combined with kinematic and anthropometric data to calculate the net internal joint moments of force at the hip, knee, and ankle using an inverse dynamics solution within The MotionMonitor software (Dempster et al., 1959; Gagnon & Gagnon, 1992).

Custom computer software (LabVIEW, National Instruments, Austin, Texas, USA) was used to generate sagittal plane hip, knee, and ankle joint power curves and individual joint EA were calculated by integrating the negative portion of each joint power curve during the 100 ms immediately following initial ground contact (vertical ground reaction force > 10 N) as described previously (McNitt-Gray, 1993; Norcross et al., 2010b). Similarly, mean internal hip extensor, knee extensor, and ankle extensor (plantarflexor) joint moments during the initial 100 ms of landing were calculated by averaging the respective net joint moment curves during periods of negative joint work (EA). The same custom software was also used to calculate the mean angular joint velocities during the initial 100 ms of landing, and to identify joint angles at initial contact and peak hip flexion, knee flexion, and ankle dorsiflexion angles between initial contact and the minimum vertical position of the whole body center of mass. We chose to isolate our analyses to the initial 100 ms of landing so that we could directly compare our results to those of Decker et al. (2003) and because ACL injury and peak strain are reported to occur during this time period (Cerulli et al., 2003; Withrow et al., 2006; Koga et al., 2010). Mean values for all dependent variables were calculated across the five trials for each landing condition. Mean joint extensor moments during periods of EA were normalized to the product of subject height and weight, while EA magnitudes were expressed as a percentage of the product of subject height and weight (% BW*Ht) to assist with presentation of the results (Norcross et al., 2013b). Positive joint moment values indicate net extensor moments for all joints, while EA values were assigned to be positive by convention to simplify their interpretation during data analysis.

Statistical analyses

Sex differences in the magnitudes of joint EA, mean joint extensor moment, and mean joint angular velocity during the initial 100 ms of landing, and peak and initial contact joint angles during the Preferred landing condition were evaluated using five separate 2 (sex) × 3 (joint) repeated measures analyses of variance (ANOVAs). Given the unexpectedly low proportion of subjects that were able to successfully complete landings in all three experimental conditions (63%), we chose to carry out the analyses of the Preferred landing condition twice, using both the total sample of subjects and the subset of successful subjects, to ensure that the preferred landing mechanics of the successful subset of subjects were similar to the preferred landing mechanics of the total sample of subjects.

For the two constrained landing conditions, the influences of sex and landing posture on the five dependent measures were evaluated using separate 2 (sex) × 2 (posture: Flexed and Erect) × 3 (joint) repeated measures ANOVAs using data collected from only the subset of subjects that successfully completed Flexed and Erect landings. When indicated by significant main or interaction effects in an ANOVA model, post-hoc mean comparisons were conducted using the Tukey–Kramer method. We specifically chose to compare the Flexed and Erect conditions independent of the Preferred condition as pilot testing indicated that the horizontal velocity of the whole body center of mass at impact was similar in the Flexed and Erect landing conditions, but slightly less than in the Preferred condition. All analyses were conducted using commercially available software (SPSS 21.0, IBM Inc., Armonk, New York, USA) with statistical significance established *a priori* as $\alpha \leq 0.05$.

RESULTS

Of the 80 subjects tested, one man and one woman were excluded from the final analyses due to software errors during data collection. While all subjects were able to hang from the drop bar and perform drop landings in the constrained conditions, 26 subjects (eight men and 18 women) were unable to contact the ground with their knee flexion angle within the desired ranges for either the Flexed (six men and nine women), Erect (two women), or both Flexed and Erect (two men and seven women) conditions. Further, three additional men were restricted from attempting drop landings in the constrained conditions due to concerns over the stability of the wooden frame to support their bodies (height = 1.93 ± 0.05 m; mass = 128.9 ± 12.5 kg). This resulted in a total sample of 78 subjects (39 women: age = 20.6 ± 2.5 years; height = 1.67 ± 0.06 m; mass = 61.4 ± 9.2 kg; 39 men: age = 21.1 ± 2.2 years; height = 1.82 ± 0.06 m;

mass = 79.8 ± 16.6 kg) who completed the preferred condition, and a subset sample of 49 subjects (21 women: age = 20.2 ± 2.0 years, height = 1.66 ± 0.06 m, mass = 60.7 ± 9.8 kg; 28 men: age = 21.4 ± 2.3 years; height = 1.81 ± 0.06 m; mass = 76.5 ± 7.5 kg) that successfully completed drop landings in all three experimental conditions.

Preferred landing condition

Overall, the results of the Preferred landing condition analyses using the subset of subjects that successfully completed all three landing tasks were generally consistent with the results using the total sample of subjects (Table 1 and Fig. 2). This indicates that excluding subjects who did not successfully complete landings in all three conditions did not result in a subset of subjects who used different preferred landing mechanics than the excluded subjects. However, it does not rule out the possibility that the subset of successful subjects is representative of a more athletic population. As a result, and to remain consistent with the analyses for the constrained landing conditions, we report only the results of the 49 subjects who successfully completed all three landing conditions.

Table 1. Preferred landing condition results

		All subjects (*n* = 78)			Successful subjects (*n* = 49)	
		Females (*n* = 39)	**Males (*n* = 39)**		**Females (*n* = 21)**	**Males (*n* = 28)**
Angle at initial contact (°)	Hip	15.7 (12.8, 18.5)	12.5 (9.7, 15.3)	Hip	15.2 (11.5, 19.0)	12.0 (8.7, 15.2)
	Knee	15.0 (12.2, 17.8)	17.7 (14.9, 20.4)	Knee	17.6 (13.8, 21.3)	16.9 (11.5, 19.0)
	Ankle[b,c]	−46.5 (−50.4, −42.7)[d]	−40.4 (−44.2, −36.6)	Ankle[b,c]	−46.1 (−51.7, −40.6)	−41.7 (−46.5, −36.8)
Peak joint flexion (°)	Hip[e]	50.0 (43.8, 56.2)	45.4 (39.3, 51.6)	Hip[e]	52.5 (43.3, 61.6)	44.5 (36.6, 52.4)
	Knee	78.5 (72.7, 84.2)	78.9 (73.2, 84.6)	Knee	81.0 (72.7, 89.3)	78.3 (71.1, 85.4)
	Ankle[b,c]	13.8 (10.7, 17.0)	14.4 (11.3, 17.6)	Ankle[b,c]	12.8 (7.8, 17.7)	14.7 (10.4, 19.0)
Mean angular velocity$_{100}$ (°/s)[f]	Hip[e,g]	263.7 (237.6, 289.7)	237.1 (211.1, 263.1)	Hip[e,g]	283.8 (245.8, 321.7)	230.6 (197.7, 263.4)
	Knee	536.9 (511.3, 562.4)	506.3 (480.8, 531.8)	Knee	541.2 (506.0, 576.4)	509.8 (479.3, 540.3)
	Ankle	510.1 (480.8, 539.4)	469.8 (440.4, 499.1)	Ankle[b]	491.0 (452.2, 529.9)	482.6 (448.9, 516.2)
Mean extensor moment$_{100}$[a]	Hip[e,g]	0.002 (−0.010, 0.014)[d]	0.021 (0.009, 0.034)	Hip[e,g]	0.020 (0.003, 0.037)	0.021 (0.006, 0.035)
	Knee	0.111 (0.102, 0.121)	0.097 (0.087, 0.106)	Knee	0.104 (0.091, 0.117)	0.100 (0.089, 0.111)
	Ankle[b]	0.053 (0.045, 0.062)	0.055 (0.046, 0.063)	Ankle[b]	0.060 (0.049, 0.072)	0.054 (0.044, 0.064)

Mean (95% CI). Hip flexion, knee flexion, and ankle dorsiflexion angles and velocities are positive (+) by convention.

[a] Normalized to body weight × height (N·m/[N·m])

[b] Ankle < knee ($P < 0.05$).

[c] Ankle < hip ($P < 0.05$).

[d] Females different than males ($P < 0.05$).

[e] Hip < knee ($P < 0.05$).

[f] Main effect for sex with females > males in all subjects only ($P < 0.05$).

[g] Hip < ankle ($P < 0.05$).

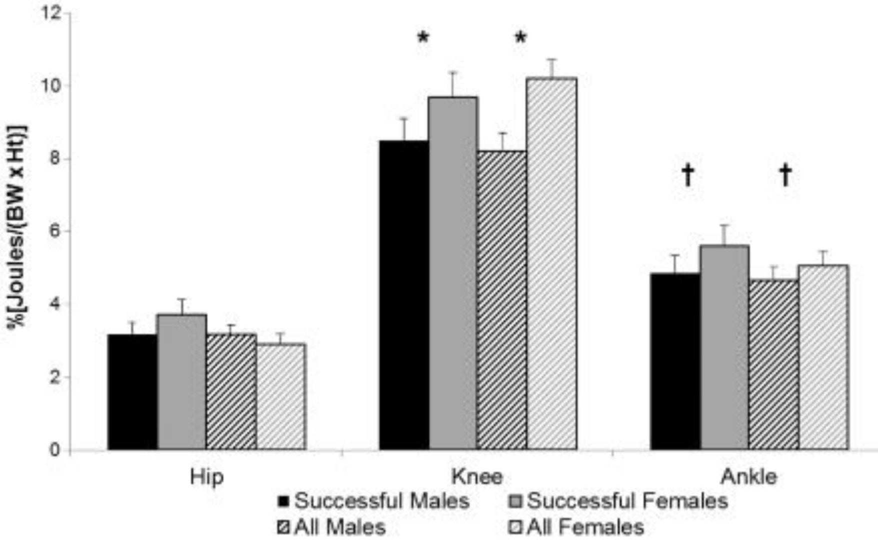

Figure 2. Influence of sex on individual joint EA during the preferred landing condition. Significantly more energy was absorbed at knee than at the ankle and hip (*) and at the ankle than at the hip (†).

During preferred landings, we failed to identify a significant sex main effect ($P = 0.939$, $\eta^2 < 0.01$) or a significant sex × joint interaction effect ($P = 0.055$, $\eta^2 = 0.06$) for joint angle at initial contact (Table 1). Similarly, no sex differences in any peak joint flexion angles during landing were identified (sex and sex × joint interaction effects, $P > 0.05$, $\eta^2 < 0.06$; Table 1). For joint EA magnitude, we identified significant main effects of sex ($P = 0.018$, $\eta^2 = 0.11$) and joint ($P < 0.001$, $\eta^2 = 0.51$). Females absorbed greater energy collapsed across joints, and the magnitude of EA was different for all joints (knee > ankle > hip; Fig. 2). However, no interaction between sex and joint was found (sex × joint interaction effect, $P = 0.788$, $\eta^2 < 0.01$; Fig. 2).

With respect to the biomechanical determinants of EA, we failed to identify a significant main effect for sex ($P = 0.413$, $\eta^2 = 0.01$) or significant sex × joint interaction effect ($P = 0.829$, $\eta^2 < 0.01$) for mean extensor moment (Table 1). Similarly, no sex differences in mean angular joint velocity were identified during landings in the Preferred condition (sex and sex × joint interaction effects, $P > 0.05$, $\eta^2 < 0.06$; Table 1). However, males and females exhibited significantly different mean extensor moment (joint main effect, $P < 0.001$, $\eta^2 = 0.57$; Table 1) and mean joint velocity (joint main effect, $P < 0.001$, $\eta^2 = 0.83$; Table 1) across joints (knee > ankle > hip) during drop landings in the Preferred condition.

Constrained landing conditions

Influence of sex

Initial contact and peak joint flexion angles for males and females during the Erect and Flexed conditions are provided in Table 2. While we identified significant sex × joint interaction effects for initial contact ($P = 0.024$, $\eta^2 = 0.08$) and peak ($P = 0.019$, $\eta^2 = 0.09$) joint flexion angles, post-hoc analyses indicated that males and females did not exhibit significantly different initial contact or peak hip, knee, or ankle joint angles, respectively (Table 2). In addition, no significant main effect for sex, or sex × posture or sex × posture × joint interactions were identified for initial contact or peak joint angles during the constrained landing conditions ($P > 0.05$, $\eta^2 \leq 0.06$; Table 2).

Table 2. Constrained landing condition results for subset of successful female ($n = 21$) and male ($n = 28$) subjects

		Erect condition			Flexed condition	
		Females	Males		Females	Males
Angle at initial contact (°)[b]	Hip	14.7 (10.1, 19.4)	13.0 (8.9, 17.0)	Hip[c]	27.9 (23.9, 32.0)	25.3 (21.8, 28.8)
	Knee	19.5 (18.6, 20.4)	19.7 (18.9, 20.6)	Knee[c]	34.0 (33.0, 35.1)	33.8 (32.9, 34.7)
	Ankle	−46.3 (−51.2, −41.5)	−38.7 (−42.7, −34.5)	Ankle[c]	−38.2 (−43.8, −32.6)	−31.2 (−36.1, −26.4)
Peak joint flexion (°)[b]	Hip	51.7 (42.4, 61.1)	41.2 (33.1, 49.3)	Hip[c]	70.9 (62.6, 79.2)	58.0 (50.8, 65.1)
	Knee	80.8 (74.8, 86.7)	74.0 (68.8, 79.2)	Knee[c]	96.1 (90.0, 102.1)	88.5 (83.2, 93.7)
	Ankle	14.3 (10.5, 18.2)	16.0 (12.7, 19.4)	Ankle[c]	18.2 (14.3, 22.2)	18.3 (14.9, 21.7)
Mean angular velocity$_{100}$ (°/s)[d]	Hip	284.8 (240.9, 328.9)	206.8 (168.7, 244.9)	Hip	290.5 (252.7, 328.3)	201.2 (168.4, 233.9)
	Knee	490.3 (452.7, 527.8)	436.4 (403.9, 468.9)	Knee[e]	441.2 (410.5, 471.9)	394.2 (367.6, 420.8)
	Ankle	494.6 (452.9, 536.4)	462.2 (426.0, 498.3)	Ankle[e]	432.3 (387.5, 477.2)	410.1 (371.3, 449.0)
Mean extensor moment$_{100}$[a,f]	Hip	0.022 (0.005, 0.039)	0.020 (0.005, 0.035)	Hip	0.028 (0.011, 0.045)	0.022 (0.007, 0.036)
	Knee	0.083 (0.071, 0.095)	0.082 (0.072, 0.093)	Knee[c]	0.089 (0.077, 0.101)	0.088 (0.077, 0.098)
	Ankle	0.061 (0.051, 0.072)	0.048 (0.039, 0.058)	Ankle	0.057 (0.047, 0.066)	0.047 (0.039, 0.055)

Mean (95% CI). Hip flexion, knee flexion, and ankle dorsiflexion angles and velocities are positive (+) by convention.

[a] Normalized to body weight × height (N·m/[N·m])

[b] Main effects for posture (flexed > erect) and joint (knee > hip > ankle) ($P < 0.05$).

[c] Posture × joint interaction with flexed > erect ($P < 0.05$).

[d] Main effects for posture (erect > flexed), joint (knee and ankle > hip), and sex (females > males) ($P < 0.05$).

[e] Posture × joint interaction with flexed < erect ($P < 0.05$).

[f] Main effects for posture (flexed > erect) and joint (knee > ankle > hip) ($P < 0.05$).

Figure 3 displays the mean EA magnitudes for males and females during Flexed and Erect landings. Females exhibited greater EA across joints and conditions than males (sex main effect, $P = 0.001$, $\eta^2 = 0.20$); but no significant sex × joint ($P = 0.824$, $\eta^2 < 0.01$), sex × posture ($P = 0.830$, $\eta^2 < 0.01$), or sex × posture × joint ($P = 0.242$, $\eta^2 = 0.03$) interactions were identified.

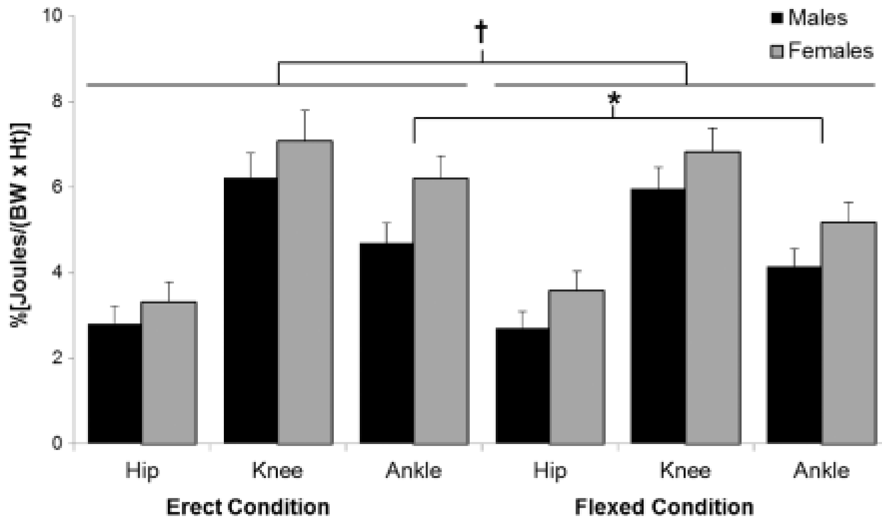

Figure 3. Influence of sex and landing posture on individual joint EA during constrained landing conditions. Significantly more energy was absorbed in the 100 ms after ground contact at the ankle (*) and across all joints (†) when landing with a flexed vs erect posture.

The results for the biomechanical determinants of EA (joint angular velocity and extensor moment) are also presented in Table 2. For both biomechanical determinants of EA, we failed to identify any significant sex × joint, sex × posture, or sex × posture × joint interactions ($P > 0.05$, $\eta^2 \leq 0.05$). The main effect of sex for mean extensor moment was also not significant ($P = 0.075$, $\eta^2 = 0.07$). However, females demonstrated significantly greater mean angular velocities than males when collapsed across joints and constrained landing conditions (sex main effect; $P = 0.002$, $\eta^2 = 0.18$).

Influence of landing posture

Peak and initial contact joint angles were significantly greater at all joints during the Flexed condition compared with the Erect condition (posture × joint interaction effects, $P < 0.001$, $\eta^2 \geq 0.92$; Table 2). Males and females exhibited greater EA across joints in the Erect condition than in the Flexed condition (posture main effect, $P < 0.001$, $\eta^2 = 0.21$), and absorbed greater energy across conditions at the knee than at the hip (joint main effect, $P < 0.001$, $\eta^2 = 0.30$; Fig. 3). Males and females both absorbed greater energy at the ankle in the Erect condition than in the Flexed condition, but no differences in the magnitude of EA at the hip or knee between constrained landing conditions were identified (posture × joint interaction, $P = 0.005$, $\eta^2 = 0.11$; Fig. 3).

For the biomechanical determinants of EA, we observed a significant posture × joint interaction effect ($P = 0.003$, $\eta^2 = 0.11$) for mean extensor moment during the 100 ms immediately after impact. Mean knee extensor moment was greater than mean ankle and hip extensor moment; and mean ankle extensor moment was greater than mean hip extensor moment during both constrained landing conditions (Table 2). Further, the magnitude of mean knee extensor moment was about 7% greater in Flexed vs Erect landings, but no significant differences in mean hip or ankle extensor moment were identified between these conditions (Table 2). Finally, mean knee and ankle flexion angular velocities were greater than hip velocities in both Flexed and Erect conditions, but knee and ankle velocities were 9% and 12% lesser, respectively, in the Flexed condition than in the Erect condition (posture × joint interaction, $P < 0.001$, $\eta^2 = 0.29$; Table 2).

DISCUSSION

The objectives of this investigation were to further investigate the potential existence of sex-specific EA strategies by evaluating the influences of sex and landing posture on joint EA and to examine the biomechanical determinants of EA to elucidate whether the mechanisms through which EA is achieved (i.e., joint angular velocity and moment) are influenced by sex and landing posture. Our primary findings are that (a) sex differences in individual joint EA are not present when the initial landing postures of males and females are similar during terminal drop landings; and (b) altering landing posture (i.e., knee flexion angle at ground contact) influences the magnitude of ankle and total lower extremity EA during the 100 ms following ground contact and the biomechanical determinants of joint EA.

Influence of sex on individual joint EA

Contrary to our hypothesis, we did not observe significant sex differences in initial contact or peak hip, knee, or ankle joint angles in healthy, recreationally active individuals performing drop landings using their preferred landing posture (Table 1). While these results are in contrast to previous research that reported that females landed with approximately 10° more ankle plantarflexion and 7° less knee flexion than males when completing the same landing task employed in our investigation (Decker et al., 2003), recent work suggests that sex differences in landing kinematics may be mitigated as the skill level of subjects increases (Bruton et al., 2013). Given the apparent difficulty of completing landings in the Flexed and Erect conditions as evidenced by the high rate of subject attrition, it is likely that the male and female subjects who were able to successfully complete landings in all three conditions were more athletic and that this increased skill level may underlie the lack of observed sex differences in initial landing posture during the Preferred condition (Table 1). Nonetheless, when utilizing similar, relatively erect postures during the Preferred condition, we did not identify sex differences in the magnitude of EA at the hip, knee, or ankle (Fig. 2). Further, the relative joint contributions to total EA (i.e., sex-specific EA strategies) were also remarkably similar as all subjects exhibited the greatest contribution to total EA from the knee, a secondary contribution from the ankle, and a tertiary contribution from the hip (Fig. 2). The lack of sex differences in individual joint EA was also evident during the two constrained landing conditions where we experimentally manipulated males and females to land with similar lower extremity configurations. Though artificially induced, our method for manipulating initial landing posture was successful, as there were no sex differences in hip, knee, or ankle joint angles at initial contact during the Erect or Flexed conditions (Table 2). Moreover, as with the Preferred condition, there were also no sex differences in the individual magnitudes of hip, knee, and ankle EA; and similar relative joint contributions to total EA when males and females performed drop landings using the same initial landing postures (Fig. 3). Collectively, we believe that our results provide sufficient evidence to conclude that sex-specific feedforward EA strategies do not exist in recreationally active adults.

Influence of landing posture on individual joint EA

In contrast to sex, initial landing posture does significantly influence the magnitude of individual joint EA; but only at the ankle. In the Erect condition, all subjects absorbed greater energy at the ankle during the initial 100 ms of landing, but comparable magnitudes of EA at the hip and knee, respectively, than during the Flexed condition (Fig. 3). Further, the relative joint contributions to total EA remained fairly consistent across conditions with mean differences (2–3%) that were much less than the 10–20% differences in hip and ankle contributions to total EA that have been previously reported between sexes (Decker et al., 2003) and following changes in landing height and technique (Zhang et al., 2000) (Fig. 3). We suggest that our methods, which did not increase the demands of the task by changing drop height, but rather simply manipulated the initial landing posture, may not have imposed enough of a perturbation to the neuromuscular system to elicit a change in the landing strategy employed (i.e., relative joint contributions to EA). However, despite a consistent EA distribution strategy during the constrained landing tasks, subjects exhibited a greater magnitude of ankle EA, and thus greater total EA across joints, during the 100 ms immediately after ground contact when landing using an erect posture. This greater magnitude of EA during the 100 ms after ground contact may be clinically relevant, as recent work indicates that greater total sagittal plane EA during this time period in individuals performing double-leg jump landings likely increases ACL loading because of sagittal plane mechanisms (Norcross et al., 2013a). However, given the inherent differences between the terminal drop landing and jump landing tasks, generalizing these findings to the current results is speculative. Regardless, it is clear that the greater total EA observed during the Erect landing condition was primarily driven by increased EA at the ankle as no significant increase in the magnitudes of hip or knee EA were identified.

Influence of sex on biomechanical determinants of EA

The second aim of this investigation was to evaluate whether the joint angular velocity and moment profiles that actually determine the magnitude of joint EA are influenced by sex and landing posture. As with joint EA magnitude, we did not identify any significant interactions between sex and joint or posture for mean extensor moment and angular velocity during the constrained landings. Further, there was not a sex main effect for mean joint extensor moment. However, we did identify a main effect for sex during the constrained conditions whereby females exhibited greater mean angular velocities than males across joints and conditions (Fig. 4). This greater angular velocity in females likely underlies the greater EA (sex main effect) noted during the constrained landing conditions (Fig. 2). Though not large enough to be statistically significant at the individual joint level, it is likely that slight increases in joint angular velocity coupled with similar joint moment profiles results in slightly greater joint EA magnitudes in females, but that these sex differences in angular velocity and EA are only statistically significant when collapsed across joints.

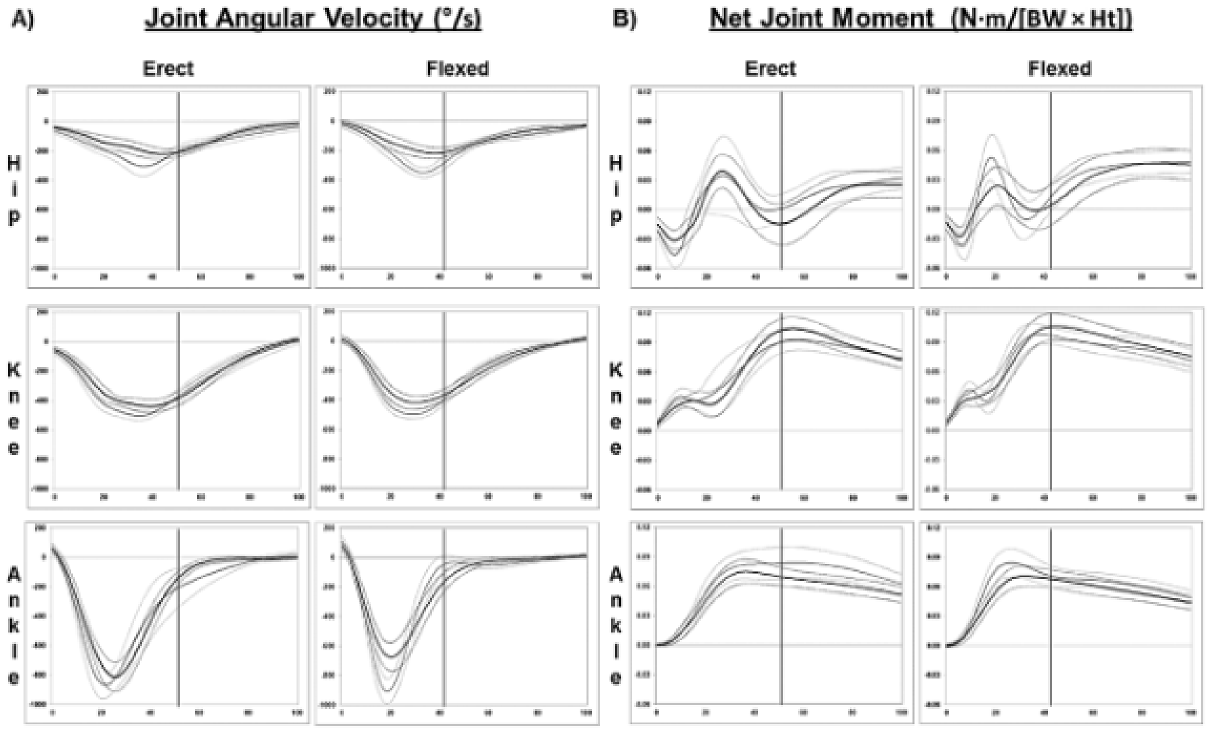

Figure 4. Ensemble (a) joint angular velocity and (b) net joint moment curves of males (black) and females (gray) performing flexed and erect landings. Means (solid lines) and 95% CIs (dashed lines) are shown with hip, knee, and ankle flexion velocities and moments (-) by convention. All data have been time-normalized from initial contact to the minimum vertical position of the whole body center of mass. Solid vertical lines indicate 100 ms after initial contact.

Influence of landing posture on biomechanical determinants of EA

Similar to joint EA magnitude, we observed that landing posture seemed to have a greater influence on joint extensor moment and velocity than sex, but that this influence was joint-specific. At the hip, mean extensor moment and angular velocity were not different during Flexed and Erect landings, which resulted in no difference in the magnitude of energy absorbed by the hip during the two constrained landing conditions. Conversely, the magnitude of ankle EA was greater during Erect landings with this increase driven by changes in ankle angular velocity as the mean ankle extensor moment did not differ during Flexed and Erect landings. Mean ankle angular velocity was 12% greater in the Erect condition than in the Flexed condition, and when combined with the similar ankle extensor moment resulted in approximately 14% greater ankle EA during Erect vs Flexed landings.

At the knee, subjects absorbed the same magnitude of energy in both Flexed and Erect conditions, but did so through different underlying mechanisms. Landings in the Flexed condition required subjects to generate approximately 7% greater mean knee extensor moment to offset about 9% lesser mean knee angular velocity than in Erect landings (Fig. 4). Collectively, these results are particularly impactful because they illustrate the delicate interplay between joint moment and joint angular velocity that combine to determine the magnitude of joint EA. While landing posture did not influence the absolute magnitude of EA absorbed at the knee, it did influence the means through which that EA was achieved. As a result, these results provide a potential explanatory mechanism for previous work that has reported more extended knee postures at initial contact during landings in fatigued vs non-fatigued conditions (Chappell et al., 2005; Benjaminse et al., 2008). During fatigued conditions, when the moment production capacity of the knee extensors is reduced, individuals might adopt a more erect landing posture that would decrease the mean knee extensor moment requirement, but allow for increased knee joint angular velocity in order to maintain the magnitude of energy absorbed by the knee. Consequently, despite reductions in the force-producing capacity of the quadriceps, the knee could remain as the primary contributor to whole body center of mass deceleration and allow for successful completion of movement tasks, even though the use of a more erect landing posture might increase the risk for injury.

Limitations

The primary limitation of this investigation is the potential that landings in the Flexed and Erect conditions were not representative of an individual's true landing performance because of the artificial manner in which we induced the desired landing postures. Given our desire to systematically manipulate and standardize initial landing postures across all subjects, it was necessary to employ a novel experimental method at the risk of potentially influencing landing performance. As a consequence of this limitation, we specifically chose not to compare the Flexed and Erect landing conditions directly to the Preferred condition, but instead opted to only compare these constrained landing conditions to each other. A second limitation is that the landing task employed (60-cm terminal drop landing) is not as closely associated with actual sporting maneuvers. However, terminal drop landing tasks have been employed in previous landing-related biomechanical studies (Decker et al., 2003; Blackburn & Padua, 2008) and the use of this task was necessary so that we could successfully manipulate and experimentally control the initial landing postures of our subjects. Further, while we chose to utilize a 60-cm drop height in order to replicate previous work (Decker et al., 2003), we can not rule out that slight differences in the relative loading between subjects may have influenced the landing strategy observed. A final limitation is the high attrition rate that was associated with the constrained landing conditions. While the subset of successful subjects displayed similar landing biomechanics to the entire sample during the Preferred condition, it is likely these subjects were more athletic than a traditional, recreationally active population.

PERSPECTIVES

When using similar lower extremity postures at initial contact, males and females did not exhibit differences in joint EA suggesting that the sex-specific EA strategies proposed by previous investigators do not exist. However, initial landing posture does influence individual joint energetics during drop landings. Compared with a Flexed landing posture, subjects absorbed greater energy at the ankle in the initial 100 ms of landing when using a more erect posture at initial contact, irrespective of sex. The increased ankle EA, driven by increased ankle joint angular velocity coupled with similar mean ankle extensor moment, contributed to greater total EA during landing. This study also elucidated an underlying mechanism through which knee joint EA can be maintained during landing by reducing mean knee extensor moment but increasing knee angular velocity during the 100 ms immediately after ground contact. In situations where knee extensor moment production is impaired such as through weakness or fatigue, an individual could adopt a more erect landing posture to allow for successful completion of a landing task (i.e., adequate EA to control motion of the whole body center of mass), but at a potential cost of placing the knee in a less favorable position relative to injury risk.

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