Gait & Posture 74 (2019) 7–13

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Full length article

Sex and limb differences during a single-leg cut with body borne load

AuraLea C. Fain, Nicholas J. Lobb, Kayla D. Seymore, Tyler N. Brown*

Dept. of Kinesiology, Boise State University, 1910 University Drive, Boise, ID, 83702, USA

ARTICLE INFO

Keywords:
Musculoskeletal injury
Kinematics
Kinetics
Military
Load carriage

ABSTRACT

Background: Military personnel don body borne loads that produce maladaptive lower limb biomechanics, increasing risk of musculoskeletal injury during common training tasks. Female personnel have over twice the injury risk as males, but it is unknown if a sex dimorphism in lower limb biomechanics exists during common training-related tasks.

Research Question: To determine whether lower limb biomechanics exhibited during a single-leg cut with military body borne loads differ between sexes.

Methods: Sixteen females and 20 males had lower limb biomechanics quantified during five single-leg cuts off each limb with four loads (20, 25, 30 and 35 kg). Each cut required participants run 4 m/s, before planting their foot on a force platform and cut 45° towards the opposite limb. Lower limb biomechanics related to musculoskeletal injury were submitted to a repeated measures ANOVA to test for main and interaction effects of load, sex, and limb.

Results: During the cut, load increased peak proximal anterior tibial shear force ($p < 0.001$) and peak hip flexion ($p = 0.010$) and knee abduction ($p = 0.045$) moments, but decreased peak knee flexion angle ($p = 0.032$). Females exhibited greater peak proximal anterior tibial shear ($p = 0.014$), and peak hip adduction ($p < 0.001$) and knee external rotation ($p = 0.001$) moment than males. Dominant limb exhibited larger peak hip adduction ($p = 0.002$); whereas, the non-dominant limb exhibited greater peak hip internal ($p = 0.002$) and knee external ($p = 0.007$) rotation moments. Only the non-dominant limb increased peak knee abduction moment ($p = 0.001$) with additional load.

Significance: During the cut, adding body borne load produced maladaptive biomechanics that may increase knee musculoskeletal injury risk. Load increased peak proximal tibial shear and potential strain of knee's soft-tissues. Females exhibited a sex dimorphism in lower limb biomechanics that may further elevate their injury risk. Both limbs exhibited biomechanics that may increase injury risk, but only the non-dominant limb further increased injury risk with load.

1. Introduction

Musculoskeletal injuries are an ever-increasing problem for the military, costing over $700 million annually [1]. These injuries commonly occur during training and cause long-term disability and medical discharge in military personnel. A majority of musculoskeletal injuries occur in the lower limb when donning body borne load during military training- related tasks, and produce pain and damage to soft-tissue structures from overuse or acute injury [2]. Body borne load routinely ranges from 20 to 40 kg during training, and produces maladaptive lower limb biomechanics that decrease physical performance and increase musculoskeletal injury risk during training [3].

During training-related tasks, such as running and cutting, body borne load increases peak vertical ground reaction force that must be attenuated by the musculoskeletal system to prevent injury [4]. To attenuate these elevated forces, military personnel may increase stance time, decrease locomotor speed and alter lower limb joint angles and moments [5]. With a 20 kg addition of body borne load, military personnel exhibit a significant reduction in peak hip and knee flexion angle when running and cutting [5,6]. These extended postures may prevent limb collapse, but aid transfer of the elevated ground reaction forces to the limb, increasing injury risk [7]. Further increasing musculoskeletal injury risk during military-related tasks are elevated joint moments produced with the addition of load. During a loaded single-leg cut, a common training- and tactical-related task [8], Brown et al. [6] reported that military personnel increase peak hip and knee flexion, and

* Corresponding author.
E-mail addresses: auraleafain@u.boisestate.edu (A.C. Fain), nicklobb@u.boisestate.edu (N.J. Lobb), kaylaseymore@boisestate.edu (K.D. Seymore), tynbrown@boisestate.edu (T.N. Brown).

https://doi.org/10.1016/j.gaitpost.2019.07.198
Received 15 January 2019; Received in revised form 8 July 2019; Accepted 16 July 2019

hip adduction joint moments to complete the maneuver. These elevated joint moments require greater force generation by the hip and knee musculature to prevent collapse of the limb, but increase stresses on the musculoskeletal system and injury risk [9]. Yet, to date, it is unknown if military personnel exhibit similar adaptations in lower limb biomechanics, and subsequent increased injury risk, when performing a single-leg cut with military-related body borne loads (20–35 kg).

During military training, female personnel are two and a half times more likely to suffer a musculoskeletal injury [1] in general and nine and half times more likely to sustain knee injury specifically than males [10]. This discrepancy in injury rates may be attributed to the sex dimorphism in lower limb biomechanics exhibited during locomotor activities, such as a single-leg cut. During unloaded cutting, females decrease peak hip and knee flexion joint angles, and increase hip adduction, and knee flexion and abduction joint moments compared to males [11]. The female's extended limb may prevent collapse of the leg, but increases injury risk by elevating peak proximal anterior tibial shear force and subsequent loading of the knee's soft-tissues, particularly the ACL [12]. Despite increased injury risk, females' lower limb biomechanics when performing a loaded single-leg cut is unknown. Previously, no significant sex differences were exhibited in lower limb biomechanics when walking with small body borne loads [13], despite the fact that females exhibit significant differences in hip and knee biomechanics during unloaded walking [14]. When running with small body borne loads, Xu et al. reported that females increase vertical ground reaction force, hip and knee joint reaction forces and tibial stresses, elevating their musculoskeletal injury risk [15]. However, it is unknown if a sex dimorphism in lower limb biomechanics exists during a single-leg cut with military- related body borne loads.

Considering policies restricting females from infantry positions, where they are required to run and cut with body borne load, were recently removed [16], it is imperative to quantify their lower limb biomechanics exhibited during single-leg cut with common military loads. This study sought to determine the lower limb biomechanics exhibited during a single-leg cut with common military body borne loads (*20, 25, 30* and *35 kg*), and compare whether biomechanics differed between sexes. We hypothesized that each addition of body borne load would produce a significant reduction in hip and knee flexion, and hip adduction joint angle, but a significant increase in hip and knee flexion, and hip adduction joint moments, and peak proximal anterior tibial shear force. We also hypothesized that females would exhibit a greater reduction in hip and knee joint angles, but larger increases hip and knee joint moments compared to males.

2. Methods

2.1. Participants

Thirty-six (20 Male and 16 Female) participants were recruited because a preliminary power analysis of peak proximal anterior tibial shear force data indicated 28 participants were needed to achieve 80% statistical power with alpha level of 0.05 (Table 1). Participants self-reported the ability to safely carry 34 kg, but were excluded if they had: (1) a history of previous back or lower extremity injury or surgery, (2) any recent injury (previous 6 months) or current pain in back or lower extremity, (3) any known neurological disorder, or (4) were currently pregnant. Research approval was acquired from the local Institutional Review Board and all participants provided written informed consent prior to testing.

Table 1
Mean (SD) subject demographics (height, mass and age) for both the male and female participants.

	N	Height (m)[a]	Mass (kg)[a]	Age (years)
Males	20	1.8 (0.1)	81.2 (10.1)	21.4 (2.8)
Females	16	1.7 (0.1)	65.9 (11.4)	21.2 (2.9)

[a] Denotes a significant main effect of sex.

2.2. Testing sessions

Each participant completed four test sessions. During each session, participants performed a single-leg cut with a different body borne load (*20, 25, 30* and *35 kg*). For each load, participants wore spandex shirt and shorts, weighted vest (Box ®, WeightVest.com, Inc., Rexburg, ID, USA), and standard-issue military helmet (ACH), and carried a mock weapon (M16) (Fig. 1). To randomize and counter-balance the test order, a 4 × 4 Latin square assigned a sequence of load conditions to each participant, prior to beginning the study. Each test session was separated by a minimum of 24 h to minimize the effects of fatigue and likelihood of injury.

2.3. Biomechanical analysis

During testing, participants had 3D lower limb biomechanical data recorded during a single-leg cut. During each cut, eight high-speed (240 fps) optical cameras (MXF20, Vicon Motion Systems, Ltd., London, UK) captured lower limb motion data, while two force platforms (OR6, Advanced Mechanical Technology, Inc., Watertown, MA.) recorded synchronous ground reaction force (GRF) data (2400 Hz). Each cut required participants to run at 4.0 m/s ± 5% through the motion capture volume before planting their foot on the force platform and performing a 45° cut towards the opposite limb. For the cut left, participants planted with their right foot and cut 45° towards the left; whereas for the cut right, participants planted with their left foot and cut 45° towards the right. Each cut direction was randomized prior to testing. For each cut, running speed was quantified from two sets of timing gates (TF100, TracTonix, Lenexa, KS, USA) placed 4 m apart immediately preceding the force platforms. Participants performed five successful cuts off each foot (right and left). For analysis each limb was defined as dominant (DOM) or non-dominant (NON) based on a footedness questionnaire (Appendix A). A successful cut required the participant cut at 45° ± 5°, only contacted the force platform with plant foot, and run the predetermined speed. During testing, participants rested between trials to minimize the effects of fatigue.

Lower limb biomechanics were quantified from 3D trajectories of 34 retro-reflective markers. Each reflective marker was attached over a specific landmark with double- sided tape and secured with elastic tape. After marker placement, each participant had stationary recording taken in anatomical position. The stationary recording was used to create a kinematic model in Visual 3D (v6, C-Motion, Inc, Germantown, MD, USA), according to our previous work [17]. The kinematic model had seven rigid segments (pelvis and bilateral foot, shank and thigh) with 27 degrees of freedom (Appendix B).

2.4. Data analysis

For each cut, synchronous GRF and 3D marker trajectories were low-pass filtered (fourth-order 12 Hz Butterworth filter). Filtered marker trajectories were processed by Visual 3D to calculate lower limb joint rotations that were expressed relative to anatomical position, using a joint coordinate systems approach. Filtered kinematic and GRF data were processed using conventional inverse dynamic analysis to obtain 3D lower limb joint moments, using inertial properties defined according to Dempster [18]. Joint moments (expressed as external) were normalized by body mass (kg) and height (m). Forces were normalized by body weight (N) and positive direction was expressed according to the corresponding orthogonal axis (i.e., peak proximal tibial shear was defined as the anteriorly directed y-axis force on the proximal tibia). All biomechanical data was time-normalized to 100% of stance phase and resampled at 1% increments (N = 101). Stance phase

Fig. 1. The equipment for each load configuration included military issue helmet (ACH), mock weapon (M16) and weighted vest. The weighted vest was systematically adjusted to provide the load necessary for each condition.

(0%–100%) was defined as heel strike to toe off (i.e., first instant the GRF exceeds and falls below 10 N, respectively).

2.5. Statistical analysis

Biomechanical variables related to musculoskeletal injury risk were selected for statistical comparison [11,14,19]. The dependent variables were peak of stance (PS, 0%–100%) hip flexion, adduction and internal rotation, and knee flexion, abduction and external rotation joint angles and moments, as well as peak proximal anterior tibial shear force. Each dependent variable was averaged across three successful trials to create a participant-based mean. Participant-based means were submitted to separate 3-way repeated measures ANOVA to test the main effects of and interaction between sex (*male, female*), load (*20, 25, 30* and *35 kg*) and limb (*DOM, NON*). Significant interactions were submitted to simple main effects and Bonferroni procedure was used for pairwise comparisons. Independent t-tests were used to compare mass and height between sexes. Effect size (ω_p2: partial omega squared) was calculated for significant main and interaction effects [20]. Statistical analysis was completed using SPSS (v23, IBM Corporation, Armonk, New York, USA) and alpha level at $p < 0.05$.

3. Results

Figs. 2 and 3 plot hip and knee biomechanics, and Tables 2 and 3 present PS values. A significant load by limb interaction was observed for PS knee abduction angle ($p = 0.018$, $\omega_p^2 = 0.07$). DOM exhibited greater PS knee abduction angle than NON with 20 kg load ($p = 0.044$); conversely, NON exhibited greater PS knee abduction angle than DOM with 25 kg load ($p = 0.039$).

A significant load by limb interaction was observed in PS knee external rotation angle ($p = 0.004$, $\omega_p2 = 0.10$). NON exhibited greater PS knee external rotation with 30 compared to 25 kg ($p = 0.041$), but significant differences were not observed between any other load.

A significant sex by limb interaction was observed for PS hip adduction angle ($p = 0.047$, $\omega_p2 = 0.09$). Further analysis revealed that females' hip adduction angle was greater than males with only NON ($p = 0.002$).

Body borne load only had significant effect on PS knee flexion angle ($p = 0.032$, $\omega_p2 = 0.16$). Participants decreased PS knee flexion with the 35 compared to 20 kg load ($p = 0.042$), but differences were not observed between other loads. Females exhibited greater PS hip adduction angle ($p = 0.006$, $\omega_p^2 = 0.19$) compared to males, but sex had no significant effect on any other joint angle. NON exhibited greater PS hip internal rotation angle ($p = 0.001$, $\omega_p2 = 0.27$) compared to DOM, however, limb did not influence any other PS joint angle.

A significant limb by sex interaction was observed for PS hip internal rotation moment ($p = 0.014$, $\omega_p2 = 0.15$). Females exhibited greater hip internal rotation moment compared to males with DOM ($p = 0.001$), but not NON. Males exhibited greater hip internal rotation moment with their NON compared to DOM ($p < 0.001$), while females exhibited no difference between limbs.

There is a significant load by limb interaction for PS knee abduction moment ($p = 0.020$, $\omega_p2 = 0.06$). NON exhibited greater PS knee abduction moment with the 35 ($p = 0.001$) and 25 ($p = 0.023$) compared to 20 kg, but differences were not observed between any other load.

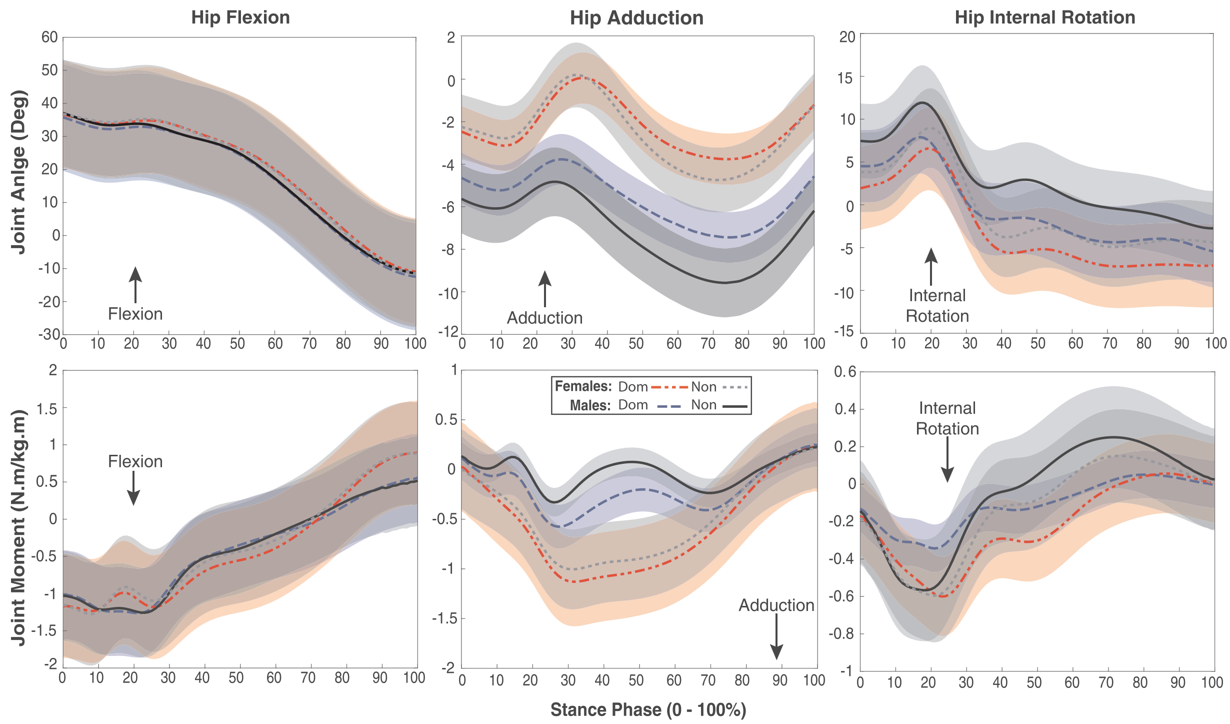

Fig. 2. Stance phase (0–100%) hip flexion, adduction and internal rotation joint angle and moments for both dominant and non-dominant limbs during the single-leg cut. Females exhibited greater peak stance hip adduction angle and moment than males. The dominant exhibited greater peak stance adduction moment than the non-dominant, while the non-dominant limb exhibited greater peak stance internal rotation moment than the dominant limb.

Body borne load had significant effect on PS hip and knee flexion ($p = 0.026$, $\omega_p2 = 0.17$ and $p = 0.027$, $\omega_p2 = 0.17$) and knee abduction ($p = 0.045$, $\omega_p2 = 0.14$) moment. Specifically, PS hip flexion and knee abduction moments were significantly greater with the 35 compared to 20 kg ($p = 0.010$ and $p = 0.015$). But, after correcting for type I error, there was no significant difference in PS knee flexion moment between any load. Females exhibited greater PS hip adduction ($p < 0.001$, $\omega_p2 = 0.41$) and knee external rotation ($p = 0.001$, $\omega_p2 = 0.29$) moments compared to males. But, sex had no significant effect on any other PS hip or knee moments. DOM exhibited significantly greater PS hip adduction moment ($p = 0.002$, $\omega_p2 = 0.24$) compared to NON, whereas, NON exhibited significantly greater PS hip

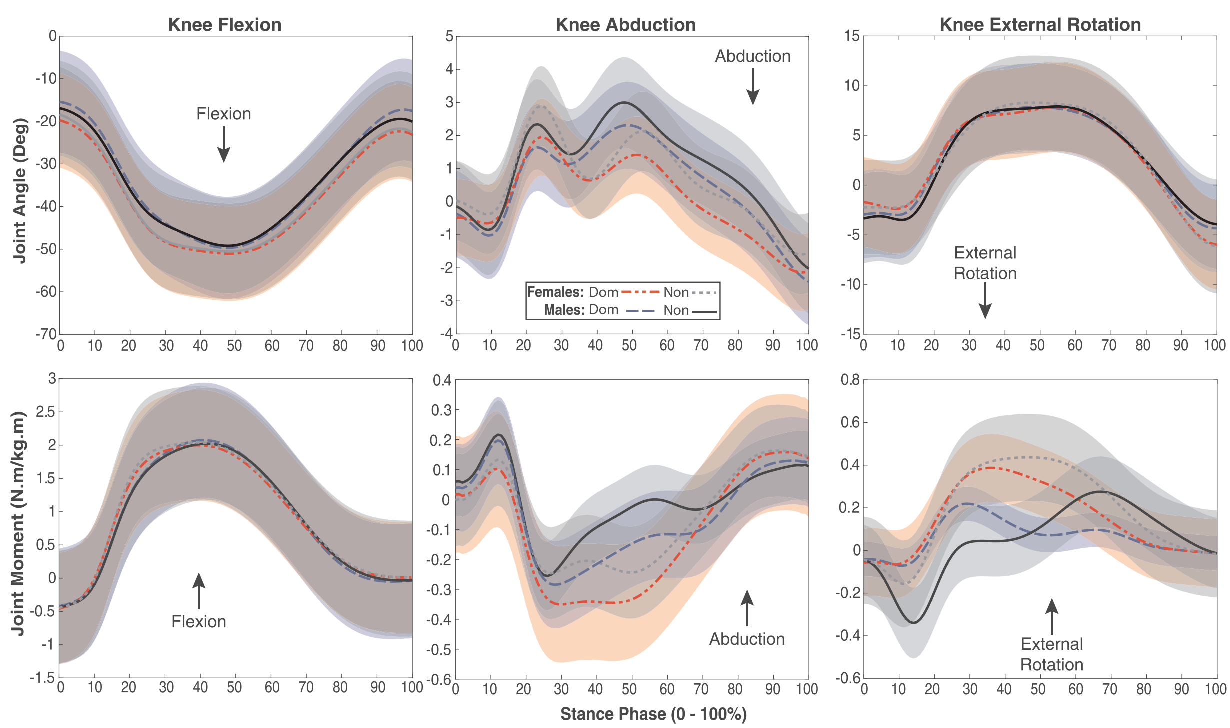

Fig. 3. Stance phase (0–100%) knee flexion, abduction and external rotation joint angles and moments for the dominant and non-dominant limbs during the single leg cut task. Both the females and the non- dominant limb exhibited greater knee external rotation moment than males and the dominant limb, respectively.

Table 2
Mean (SD) peak stance (0%–100%) hip and knee joint angles for both the dominant and non-dominant limbs exhibited by the male and female participants during the single-leg cut with each body borne load.

		20kg		25 kg		30 kg		35 kg	
		Dom	Non	Dom	Non	Dom	Non	Dom	Non
Hip Flx	M	36.7 (7.2)	36.9 (6.1)	37.2 (6.7)	38.0 (5.6)	35.8 (6.3)	37.1 (5.5)	36.5 (6.6)	38.2 (6.9)
	F	38.3 (6.8)	38.4 (4.6)	36.6 (5.1)	37.8 (4.8)	37.3 (7.6)	37.0 (5.5)	39.3 (7.3)	38.1 (5.8)
Hip Add[a,d]	M	−0.2 (4.1)	−2.1 (4.0)	−0.1 (5.4)	−1.6 (5.4)	−0.3 (4.2)	−3.4 (4.5)	−0.9 (4.0)	−1.5 (4.3)
	F	1.7 (4.6)	2.2 (5.8)	3.1 (5.3)	3.7 (5.7)	2.2 (4.1)	−2.4 (4.3)	2.2 (5.7)	2.4 (4.7)
Hip Int Rot[b]	M	9.4 (7.7)	13.3 (8.0)	8.9 (6.8)	13.3 (7.1)	10.6 (8.0)	14.3 (6.8)	8.2 (8.8)	12.1 (6.5)
	F	7.7 (5.2)	11.1 (9.5)	8.3 (6.5)	9.0 (7.1)	7.3 (4.6)	11.0 (4.9)	8.4 (6.6)	10.1 (5.4)
Knee Flx[c]	M	−50.4 (8.9)	−49.4 (8.2)	−50.0 (6.3)	−49.8 (6.2)	−50.1 (7.8)	−50.1 (7.6)	−49.8 (7.3)	−49.0 (6.1)
	F	−53.7 (6.0)	−53.4 (5.1)	−51.8 (6.4)	−51.6 (7.4)	−51.6 (6.9)	−50.5 (6.0)	−50.6 (7.3)	−49.8 (5.5)
Knee Abd[e]	M	−3.5 (3.0)	−2.3 (2.0)	−2.7 (2.7)	−3.0 (2.9)	−3.8 (2.9)	−2.9 (2.6)	−3.0 (2.1)	−2.6 (3.0)
	F	−3.4 (3.5)	−2.3 (3.3)	−1.7 (2.1)	−3.6 (3.5)	−3.3 (3.1)	−2.6 (3.5)	−3.9 (3.3)	−3.1 (1.7)
Knee Ext Rot[e]	M	−5.6 (2.8)	−6.5 (2.8)	−6.1 (2.7)	−6.7 (3.5)	−6.1 (3.9)	−5.3 (3.3)	−6.5 (3.2)	−6.6 (3.6)
	F	−6.7 (4.4)	−7.9 (4.1)	−6.9 (2.5)	−7.9 (3.0)	−7.7 (3.0)	−6.7 (2.8)	−7.0 (3.2)	−7.0 (3.1)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by limb interaction.
[e] Denotes a significant limb by load interaction.

internal rotation ($p = 0.002$, $\omega_p2 = 0.24$) and knee external rotation ($p = 0.007$, $\omega_p2 = 0.18$) moments compared to DOM. Limb had no effect any other hip or knee moment.

Body borne load had significant effect on peak proximal anterior tibial shear ($p < 0.001$, $\omega_p2 = 0.56$), which was larger with 35 and 30 compared to 20 kg ($p < 0.001$ and $p = 0.009$) and 35 compared to 25 ($p = 0.019$) kg load. No significant differences were observed between any of the other loads. Females exhibited significantly larger peak proximal anterior tibial shear than males ($p = 0.014$, $\omega_p2 = 0.15$). Limb had no effect on tibial shear.

4. Discussion

Adding body borne load produced maladaptive biomechanics that may increase knee musculoskeletal injury risk. During the cut, participants increased peak proximal anterior tibial shear between 16% and 27% when adding 5 to 15 kg's of load. Large anteriorly directed proximal tibial shear forces reportedly load the knee's soft-tissue structures that restrain anterior tibial translation (i.e. ACL) [12]. Consequently, this shear force is an indicator of ACL loading and could be considered an injury risk factor – particularly for females. In agreement with Chappell et al. [9], females exhibited 54% greater peak proximal anterior tibial shear than the males. These "hazardous" knee forces may subsequently increase females' injury risk and stem from neuromuscular control differences. During dynamic tasks, such as the single-leg cut, females exhibit a quadriceps-dominant neuromuscular strategy, which is characterized by heavy reliance upon the knee extensors to successfully complete the task [21]. A significant increase in knee extensor activity is exhibited during load carriage [13] and purportedly increases proximal anterior tibial shear force [22]. Females currently presented 28% weaker knee extensors than males (Appendix C), potentially elevating quadriceps activation and leading to "hazardous" knee shear forces during the loaded cuts. However, it is unknown if female military personnel are more reliant upon a quadriceps-dominant neuromuscular strategy, further increasing their knee injury risk to complete loaded cuts.

Table 3
Mean (SD) peak stance (0%–100%) hip and knee joint moments and proximal anterior tibial shear shear for both the dominant and non-dominant limbs exhibited by the male and female participants during the single-leg cut with each body borne load.

		20kg		25 kg		30 kg		35 kg	
		Dom	Non	Dom	Non	Dom	Non	Dom	Non
Hip Flx[c]	M	−1.5 (0.4)	−1.4 (0.3)	−1.7 (0.6)	−1.6 (0.5)	−1.6 (0.5)	−1.6 (0.4)	−1.6 (0.6)	−1.6 (0.4)
	F	−1.5 (0.5)	−1.4 (0.4)	−1.5 (0.5)	−1.5 (0.5)	−1.5 (0.5)	−1.5 (0.5)	−1.6 (0.5)	−1.6 (0.6)
Hip Add[a,b]	M	−0.8 (0.4)	−0.5 (0.3)	−0.8 (0.4)	−0.6 (0.3)	−0.8 (0.4)	−0.6 (0.3)	−0.8 (0.4)	−0.6 (0.3)
	F	−1.2 (0.4)	−1.1 (0.4)	−1.3 (0.4)	−1.1 (0.4)	−1.2 (0.4)	−1.1 (0.4)	−1.2 (0.4)	−1.1 (0.4)
Hip Int Rot[b,d]	M	−0.4 (0.2)	−0.6 (0.3)	−0.5 (0.2)	−0.6 (0.6)	−0.5 (0.2)	−0.7 (0.3)	−0.5 (0.2)	−0.7 (0.3)
	F	−0.7 (0.3)	−0.8 (0.4)	−0.6 (0.2)	−0.6 (0.3)	−0.7 (0.3)	−0.7 (0.2)	−0.7 (0.2)	−0.7 (0.3)
Knee Flx[c]	M	2.0 (0.4)	2.0 (0.3)	2.2 (0.5)	2.2 (0.3)	2.2 (0.4)	2.1 (0.4)	2.2 (0.3)	2.1 (0.4)
	F	2.0 (0.2)	2.1 (0.3)	2.1 (0.4)	2.1 (0.4)	2.0 (0.3)	2.1 (0.3)	2.1 (0.3)	2.1 (0.4)
Knee Abd[c,e]	M	0.3 (0.2)	0.3 (0.2)	0.3 (0.2)	0.4 (0.3)	0.4 (0.3)	0.4 (0.2)	0.4 (0.2)	0.4 (0.2)
	F	0.3 (0.2)	0.3 (0.1)	0.3 (0.2)	0.3 (0.2)	0.3 (0.2)	0.3 (0.1)	0.3 (0.2)	0.4 (0.1)
Knee Ext Rot[a,b]	M	0.3 (0.2)	0.4 (0.2)	0.3 (0.1)	0.4 (0.3)	0.3 (0.2)	0.3 (0.2)	0.3 (0.1)	0.4 (0.2)
	F	0.4 (0.2)	0.6 (0.2)	0.5 (0.2)	0.6 (0.3)	0.4 (0.2)	0.5 (0.2)	0.4 (0.2)	0.5 (0.3)
Prox Tib. Shear	M	0.1 (0.1)	0.1 (0.1)	0.1 (0.1)	0.1 (0.1)	0.1 (0.1)	0.1 (0.1)	0.1 (0.1)	0.1 (0.1)
	F	0.2 (0.1)	0.1 (0.1)	0.2 (0.1)	0.2 (0.1)	0.2 (0.1)	0.2 (0.1)	0.2 (0.1)	0.2 (0.1)

[a] Denotes a significant main effect of sex.
[b] Denotes a significant main effect of limb.
[c] Denotes a significant main effect of load.
[d] Denotes a significant sex by load interaction.
[e] Denotes a significant limb by load interaction.

To successfully execute the loaded single-leg cut, participants exhibited hip and knee biomechanics thought to increase injury risk and decrease physical performance. In agreement with previous work [6], participants decreased peak knee flexion with the 15 kg addition of load. An extended knee is reportedly characteristic of a stiff limb [4]. The current participants decreased peak knee flexion by 10° compared to unloaded cuts [23], which may stiffen the limb and aid the musculature in decelerating the center of mass during the loaded cuts [4]. But, the stiff limb may strain the joint's soft-tissue structures and increase injury risk. Participants may further elevate knee injury risk by increasing knee abduction moment by 22% when increasing load from 20 to 35 kg. Large knee abduction moments are a predictor for knee injury [24], and may substantially elevate injury risk. In contrast to previous literature [6] and our hypothesis, reduced peak hip flexion was not observed with the addition of load; yet, current participants reduced peak hip flexion by 15° compared to unloaded cuts [25]. Moreover, participants exhibited significant 10% increase in peak hip flexion moment with the 35 kg load. Large, proximal hip musculature provides propulsion and stabilizes the trunk during the cutting task [26]. The heavier trunk may require participants use the larger, stronger hip muscles to successfully complete the cut, but impairs the individual's ability to perform the maneuver [6].

Females exhibited frontal and transverse plane hip and knee biomechanics related to knee musculoskeletal injury during the loaded cut. In agreement with previous literature, females used larger hip adduction angle and moment, and hip internal and knee external rotation moments compared to males [11,27]. Although joint moments are closely related to activation around a joint [28], they fail to directly measure muscular forces and should be treated as indicators of joint loading, rather than mechanisms of injury. Regardless the hip and knee biomechanics adopted by the females purportedly strain knee structures associated with stabilization [24], and may increase ACL injury risk. Large knee external rotation moments paired with anterior tibial shear force reportedly further increases ACL loading and may present a worst-case scenario in terms of injury risk during loaded cuts [19]. Although, the reason for the current sex dimorphism is not immediately evident, it may be attributed to strength differences between male and female participants. Females' hip and knee musculature was between 14% and 28% weaker than males (Appendix C). Thus, females may not possess the strength to prevent "hazardous" hip and knee biomechanics to complete the loaded single-leg cut.

Both limbs exhibited hip and knee biomechanics reported to increase injury risk. Previous literature is inconclusive on whether substantial differences in biomechanics exist between limbs, yet current outcomes support a significant limb dimorphism during the loaded single-leg cuts. Specifically, DOM exhibited 0.176 Nm/kgm greater peak hip adduction moment than NON; whereas, NON exhibited greater hip internal rotation angle and moment, and knee external rotation than DOM. The DOM hip abductors are reportedly stronger than NON [29]. This may afford DOM greater hip neuromuscular control, as evidence by the larger hip adduction moment, to execute the cut. Conversely, NON may further increase knee injury risk during the loaded cut by exhibiting significant 29%–36% increase in peak knee abduction moment with addition of the chosen loads [24]. These large knee abduction moments may elevate the NON's knee injury risk. In fact, ACL injured individuals exhibit smaller peak knee abduction moments than currently observed in NON (0.30 vs 0.36 Nm/kgm, respectively) [30]. However, the incidence rate of training-related musculoskeletal injury between limbs is unknown and warrants further exploration.

The chosen participants may be a limitation. All participants self-reported the ability to safely carry a 34 kg body borne load, but were not required to have load carriage experience. Participants who routinely carry body borne loads, such as military personnel, may be stronger, leading to altered biomechanics compared to inexperienced and/or weaker load carriers. Although, we are currently unaware of differences in lower limb biomechanics exhibited by experienced and inexperienced load carriers, it may be experience and strength, rather than sex, impact hip and knee biomechanics during training-related locomotor tasks.

5. Conclusion

In conclusion, body borne load produced maladaptive lower limb biomechanics that may increase knee injury risk during a single-leg cut. With load, participants increased anteriorly directed peak proximal tibial shear force and strain of knee's soft-tissues. Females exhibited a sex dimorphism in lower limb biomechanics that may further elevate their injury risk. Compared to the stronger males, females exhibited a larger peak proximal tibial shear force, and frontal and transverse plane hip and knee biomechanics related to knee injury to complete the cut. During the single- leg cut, both DOM and NON exhibited lower limb biomechanics that may increase knee injury risk, such as DOM's greater hip adduction moment or NON's greater hip internal rotation and knee external rotation moments. But, load further increased peak knee abduction moment, and subsequent risk of soft-tissue injury, in only the non- dominant limb.

Declaration of Competing Interest

None of the authors demonstrate any conflict of interest regarding this submission.

Acknowledgements

This work was supported by was Battelle Energy Alliance/Idaho National Laboratory (https://www.inl.gov/) and Natick Soldier Research Development and Engineering Center (https://www.nsrdec.army.mil/) under award number: DE-AC07- 05ID14517.

Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:https://doi.org/10.1016/j.gaitpost.2019.07.198.

References

[1] B.C. Nindl, T.J. Williams, P.A. Deuster, N.L. Butler, B.H. Jones, Strategies for optimizing military physical readiness and preventing musculoskeletal injuries in the 21st century, US Army Med. Dep. J. 13 (2013) 5–23.

[2] K.G. Hauret, B.H. Jones, S.H. Bullock, M. Canham-Chervak, S. Canada, Musculoskeletal injuries: Description of an under-recognized injury problem among military personnel, Am. J. Prev. Med. (2010), https://doi.org/10.1016/j.amepre.2009.10.021.

[3] K.A. Andersen, P.N. Grimshaw, R.M. Kelso, D.J. Bentley, Musculoskeletal lower limb injury risk in army populations, Sport. Exerc. Med. - Open J. 2 (2016) 22, https://doi.org/10.1186/s40798-016-0046-z.

[4] J.W. Ramsay, C.L. Hancock, M.P. O'Donovan, T.N. Brown, Soldier-relevant body borne loads increase knee joint contact force during a run-to-stop maneuver, J. Biomech. (2016), https://doi.org/10.1016/j.jbiomech.2016.10.022.

[5] T.N. Brown, M. O'Donovan, L. Hasselquist, B.D. Corner, J.M. Schiffman, Body borne loads impact walk-to-run and running biomechanics, Gait Posture (2014), https://doi.org/10.1016/j.gaitpost.2014.04.001.

[6] T.N. Brown, M. O'Donovan, L. Hasselquist, B. Corner, J.M. Schiffman, Soldier-relevant loads impact lower limb biomechanics during anticipated and unanticipated single-leg cutting movements, J. Biomech. (2014), https://doi.org/10.1016/j.jbiomech.2014.09.002.

[7] J.L. McNitt-Gray, Kinetics of the lower-extremities during drop landings from 3 heights, J. Biomech. 26 (1993) 1037–1046, https://doi.org/10.1016/S0021-9290(05)80003-X.

[8] E.A. Harman, D.J. Gutekunst, P.N. Frykman, M.A. Sharp, B.C. Nindl, J.A. Alemany, R.P. Mello, Prediction of simulated battlefield physical performance from field-expedient tests, Mil. Med. (2008), https://doi.org/10.7205/MILMED.173.1.36.

[9] J.D. Chappell, B. Yu, D.T. Kirkendall, W.E. Garrett, A comparison of knee kinetics between male and female recreational athletes in stop-jump tasks, Am. J. Sports Med. 30 (2002) 261–267, https://doi.org/10.1177/03635465020300021901.

[10] K.E. Johnson, Tears of cruciate ligaments of the knee, US armed forces, 1990-2002, Med. Surveill. Mon. Rep. 9 (2003) 2–6.

[11] C.D. Pollard, S.M. Sigward, C.M. Powers, Limited hip and knee flexion during landing is associated with increased frontal plane knee motion and moments, Clin.

Biomech. Bristol Avon (Bristol, Avon) 25 (2010) 142–146, https://doi.org/10.1016/j.clinbiomech.2009.10.005.
[12] K.L. Markolf, D.M. Burchfield, M.M. Shapiro, M.F. Shepard, G.A.M. Finerman, J.L. Slauterbeck, Combined knee loading states that generate high anterior cruciate ligament forces, J. Orthop. Res. 13 (1995) 930–935, https://doi.org/10.1002/jor.1100130618.
[13] A. Silder, S.L. Delp, T. Besier, Men and women adopt similar walking mechanics and muscle activation patterns during load carriage, J. Biomech. 46 (2013) 2522–2528, https://doi.org/10.1016/j.jbiomech.2013.06.020.
[14] S.H. Cho, J.M. Park, O.Y. Kwon, Gender differences in three dimensional gait analysis data from 98 healthy Korean adults, Clin. Biomech. Bristol Avon (Bristol, Avon) 19 (2004) 145–152, https://doi.org/10.1016/j.clinbiomech.2003.10.003.
[15] C. Xu, A. Silder, J. Zhang, J. Reifman, G. Unnikrishnan, A cross-sectional study of the effects of load carriage on running characteristics and tibial mechanical stress: implications for stress-fracture injuries in women, BMC Musculoskelet. Disord. 18 (2017) 125, https://doi.org/10.1186/s12891-017-1481-9.
[16] B. Chappell, Pentagon says women can now serve in front-line ground combat positions, Natl. Public Radio. (2015).
[17] T.N. Brown, J.T. Kaplan, S.E. Cameron, K.D. Seymore, J.W. Ramsay, Individuals with varus thrust do not increase knee adduction when running with body borne load, J. Biomech. (2018), https://doi.org/10.1016/j.jbiomech.2018.01.003.
[18] W.T. Dempster, Space requirements of the seated operator, Am. J. Phys. Anthro. 22 (1955) 1–254 doi:citeulike-article-id:437004.
[19] T.C. Sell, J.S. Akins, A.R. Opp, S.M. Lephart, Relationship between tibial acceleration and proximal anterior tibia shear force across increasing jump distance, J. Appl. Biomech. 30 (2014) 75–81, https://doi.org/10.1123/jab.2012-0186.
[20] G. Keren, C. Lewis, Partial omega squared for anova designs, Educ. Psychol. Meas. (1979), https://doi.org/10.1177/001316447903900116.
[21] S.M. Sigward, C.M. Powers, The influence of gender on knee kinematics, kinetics and muscle activation patterns during side-step cutting, Clin. Biomech. Bristol Avon (Bristol, Avon) 21 (2006) 41–48, https://doi.org/10.1016/j.clinbiomech.2005.08.001.
[22] T.N. Brown, S.G. McLean, R.M. Palmieri-Smith, Associations between lower limb muscle activation strategies and resultant multi-planar knee kinetics during single leg landings, J. Sci. Med. Sport 17 (2014) 408–413, https://doi.org/10.1016/j.jsams.2013.05.010.
[23] S.G. McLean, X. Huang, A. Su, A.J. Van Den Bogert, Sagittal plane biomechanics cannot injure the ACL during sidestep cutting, Clin. Biomech. Bristol Avon (Bristol, Avon) 19 (2004) 828–838, https://doi.org/10.1016/j.clinbiomech.2004.06.006.
[24] T.E. Hewett, G.D. Myer, K.R. Ford, R.S. Heidt, A.J. Colosimo, S.G. McLean, A.J. van den Bogert, M.V. Paterno, P. Succop, Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes, Am. J. Sports Med. 33 (2005) 492–501, https://doi.org/10.1177/0363546504269591.
[25] M.S. Chan, C.F. Huang, J.H. Chang, T.W. Kernozek, Kinematics and kinetics of knee and hip position of female basketball players during side-step cutting with and without dribbling, J. Med. Biol. Eng. (2009).
[26] R.R. Neptune, K. Sasaki, S.A. Kautz, The effect of walking speed on muscle function and mechanical energetics, Gait Posture 28 (2008) 135–143, https://doi.org/10.1016/j.gaitpost.2007.11.004.
[27] C.D. Pollard, I.M.C. Davis, J. Hamill, Influence of gender on hip and knee mechanics during a randomly cued cutting maneuver, Clin. Biomech. Bristol Avon (Bristol, Avon) (2004), https://doi.org/10.1016/j.clinbiomech.2004.07.007.
[28] D.G. Lloyd, T.F. Besier, An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo, J. Biomech. (2003), https://doi.org/10.1016/S0021-9290(03)00010-1.
[29] C. Jacobs, T.L. Uhl, M. Seeley, W. Sterling, L. Goodrich, Strength and fatigability of the dominant and nondominant hip abductors, J. Athl. Train. 40 (2005) 203–206.
[30] S.B. Clarke, I.C. Kenny, A.J. Harrison, Dynamic knee joint mechanics after anterior cruciate ligament reconstruction, Med. Sci. Sports Exerc. (2015), https://doi.org/10.1249/MSS.0000000000000389.