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Females and males use different hip and knee mechanics in response to symmetric military-relevant loads

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ABSTRACT

The purpose of this study was to determine if females and males use different hip and knee mechanics when walking with standardized military-relevant symmetric loads. Fifteen females and fifteen males walked on a treadmill for 2-min at a constant speed under three symmetric load conditions (unloaded: 1.71 kg, medium: 15 kg, heavy: 26 kg). Kinematic and kinetics of the hip and knee were calculated in the sagittal and frontal planes of the dominant limb. In females, hip abduction moments (normalized to total mass) and sagittal knee excursion decreased with increased load ($p \leq 0.024$). In males, hip frontal excursion and adduction angle increased with load ($p \leq 0.003$). Females had greater peak hip adduction angle than males in the unloaded and medium load conditions ($p \leq 0.036$). Across sex, sagittal hip and knee excursion, peak knee extension angle, and peak hip and knee flexion angles increased with increased load ($p \leq 0.005$). When normalized to body mass, all peak joint moments increased with each load ($p \leq 0.016$) except peak hip adduction moment. When normalized to total mass, peak hip adduction moment and knee flexion, extension, and adduction moments decreased with each load ($p < 0.001$). While hip frontal plane kinetic alterations to load were only noted in females, kinematic changes were noted in males at the hip and females at the knee. Differences in strategies may increase the risk of hip and knee injuries in females compared to males. This study noted load and sex effects that were previously undetected, highlighting the importance of using military-relevant standardized loads and investigating frontal plane adaptations.

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1. Introduction

Load carriage is an essential task of combat Soldiers (Nindl et al., 2016) and is associated with musculoskeletal injury (Knapik and Reynolds, 2015; Schuh-Renner et al., 2017). Despite females having access to all combat-centric occupations, the majority of military-relevant research investigating how load affects gait mechanics focuses on male participants (Harman et al., 2000; Lenton et al., 2019; Seay et al., 2014), and demonstrates that gait is altered by load magnitude (Attwells et al., 2006; Birrell and Haslam, 2009) and distribution (Birrell and Haslam, 2010; Majumdar et al., 2010), and by walking speed (Harman et al., 2000; LaFiandra et al., 2003; Lenton et al., 2019). While testing loads based on an individual's body mass reduces the effect of differences in body size, Soldiers do not adjust loads based on body mass and bring what is required for the mission (Dean and DuPont, 2004; Department of the Army, 1990). Load distribution affects gait mechanics, especially in the sagittal plane (Attwells et al., 2006; Lenton et al., 2019; Seay et al., 2014). Prior military load carriage research has focused on posterior loads (Harman et al., 2000; Kinoshita, 1985; Liew et al., 2016), but reports from current operations suggest that symmetric loads are increasing due to equipment being distributed anteriorly on torso-born body armor rather than in a backpack alone (Dean and DuPont, 2004; Hasselquist et al., 2018; Seay et al., 2014).

Female Soldiers have more than twice the risk of any musculoskeletal injury compared to their male counterparts (Jones et al., 2017) and are at substantially higher risk of femoral neck (Kupferer et al., 2014) and pubic ramus (Hill et al., 1996) stress fractures and patellofemoral pain (PFP) (Boling et al., 2010). Previous research has focused on PFP due to its high prevelance in females. However, in the military, stress fractures of the hip region are considered among the most severe injuries (Lee et al., 2003;

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Scott et al., 2012) despite their low prevelance. Altered frontal plane mechanics of both the hip and knee may be linked to PFP (Powers, 2010) while hip mechanics may be linked to femoral neck (Edwards et al., 2008) and pubic rami stress fractures (Kelly et al., 2000).

Despite known sex-differences in hip and knee mechanics during unloaded walking (Bruening et al., 2015; Kerrigan et al., 1998), military load carriage research has not noted these differences (Krupenevich et al., 2015; Silder et al., 2013). The analysis in military-relevant load carriage research both in males (Attwells et al., 2006; Harman et al., 2000; Lenton et al., 2019; Seay, 2015; Seay et al., 2014) and when comparing sexes (Krupenevich et al., 2015; Silder et al., 2013) has been limited to the sagittal plane. The distribution and magnitude of loads in previous sex comparison studies may have obscured sagittal plane differences. Sex-specific changes in frontal plane hip and knee mechanics during military-relevant load carriage may be important for understanding hip and knee injury mechanisms.

Given that females have access to combat-centric occupations requiring load carriage and are at increased risk of musculoskeletal injury, research is needed to investigate sagittal and frontal plane gait mechanics in females and males when carrying military-relevant loads. The purpose of this study was to determine if females and males use different sagittal and frontal plane hip and knee kinematics and kinetics when walking with standardized military-relevant symmetric loads. We hypothesized that females would respond differently to increased load than males even after accounting for body mass.

2. Methods

2.1. Participants

Fifteen females and fifteen males were recruited from the Boston area and consisted of civilians, Reserve Officer Training Corp (ROTC) cadets, and active-duty military (Table 1). We sought to recruit individuals with experience carrying load and who were unlikely to become fatigued during testing. Therefore, participants needed to be between 18 and 40 years of age, and report either having experience walking with a 9.09 kg load or greater for at least 30 min or strenuously exercising at least 3-times per week. Exclusion criteria included having an acute or chronic injury that affected walking ability, exercise routine, or load carrying ability. This study was approved by the Boston University Institutional Review Board. Participants provided written informed consent prior to participation.

Table 1
Participant descriptive statistics.

Anthropometrics (mean ± SD)	Female (n = 15)	Male (n = 15)	*p-value*[a]
Age (yrs)	26.1 ± 5.1	26.0 ± 4.8	0.971
Height (m)	1.65 ± 0.08	1.78 ± 0.08	**<0.001**
Mass (kg)	67.8 ± 9.4	79.8 ± 9.1	**0.001**
%Body Fat	23.8 ± 3.4%	10.6 ± 3.7%	**0.001**
BMI	24.8 ± 2.4	25.3 ± 1.7	0.561
Demographics	*N*	*N*	
Military			
Cadet	3	3	–
Active Duty	–	1	–
Recreational Hiker	10	8	–
Load Carriage Experience	12	12	–
Exercise 3/week	13	14	–
Dominant Foot			
Right	14	12	–
Left	1	3	–
Non-parametric tests	*Median (Range)*	*Median (Range)*	*p-value*[b]
Activity Score	10.00 (5–10)	10.00 (7–10)	0.116
RPE[*]			
Unloaded	8 (7–9)	8 (7–10)	0.713
Medium Load	9 (7–10)	9 (7–11)	0.967
Heavy Load	10 (8–12)	10 (8–13)	0.775
Loads relative to body mass			
Medium Load (%BM, mean ± SD)	19.3 ± 2.6%	16.8 ± 2.4%	–
Range	(14.8%–23.0%)	(13.0%–21.8%)	–
Heavy Load (%BM, mean ± SD)	36.8 ± 4.8%	30.6 ± 3.1%	–
Range	(28.4%–43.8%)	(24.9%–35.9%)	–
%Change (mean ± SD)	17.6 ± 2.2%	13.8 ± 1.6%	–

Note: Bold values indicate significance $p < 0.05$.
[a] Indicates Independent *T*-test.
[b] Indicates Mann-Whitney *U* test.
[*] Borg RPE 6-20 Scale: '6' refers to 'Zero Exertion' and '20' refers to 'Maximal Exertion'.

2.2. Instrumentation

We used a 10-camera motion capture system (Vicon Motion Systems Ltd, Centennial, CO) to collect marker position data (100 Hz). Thirty reflective markers and four rigid plastic marker clusters quantified trunk, pelvis, and lower extremity kinematics and lower extremity kinetics (Ogamba et al., 2016). An instrumented split-belt treadmill (Bertec Corporation, Columbus, OH) measured ground reaction force (GRF) data (2000 Hz).

2.3. Conditions

A symmetrically loaded weight vest (V-Force Weightvest.com Inc. Rexburg, ID; Fig. 1) was used to simulate three load conditions during walking. The loads for each condition were 1) unloaded, weight vest only (∼1.7 kg); 2) medium load (∼15 kg); and 3) heavy load (∼26 kg), and were presented block randomized to reduce order effects. The loads mimicked a recommended fighting load (22 kg) and approach march load (33 kg) (Department of the Army, 1990), minus mass from other equipment not carried on the torso (i.e., helmet, boots, and weapon; ∼7.3 kg) (Loverro et al., 2015). Symmetric loads better reflect distributions used in current operations with equipment distributed to the torso-born body armor rather than in a backpack (Dean and DuPont, 2004; Seay et al., 2014). Consistent with Seay et al. (2014), using symmetric loads reduces the effect of changing the whole system center of mass (COM) location in the anteroposterior direction allowing consequences of load magnitude to be examined rather than changes in COM location. A walking speed of 1.35 m/s, the recommended standard foot march rate (Department of the Army, 1990), was used to control speed effects (Harman et al., 2000; LaFiandra et al., 2003).

2.4. Data acquisition

Participants completed questionnaires including the UCLA Activity Score (Terwee et al., 2011) and demographic information. Body fat composition was determined using an American College of Sports Medicine recommended 3-site skinfold technique (female: triceps, abdominal, and supra-iliac crest; male: triceps, chest, and subscapular) (Jackson and Pollock, 1985).

Participants wore the weight-vest, tight-fitting shorts, shirt, and their own exercise shoes. For each condition, participants became accustomed to walking with the weight-vest and load by walking a 25-m loop 5-times. For each condition, a static standing trial was collected, followed by the walking trial. Two minutes of continuous walking data were recorded once the treadmill reached 1.35 m/s. Participants were asked every minute to rate their per-

Fig 1. Female (a) and male (b) with medium and heavy load condition.

ceived exertion (RPE) (Borg, 1982) to monitor fatigue and were given at least two minutes of rest between each trial to reduce the chance of fatigue.

2.5. Data processing

Marker and GRF data were filtered in Visual3D (v6, C-Motion, Inc. Germantown, MD) using a low-pass, fourth-order Butterworth filter with cutoff frequencies of 6 Hz and 10 Hz, respectively (Ogamba et al., 2016; Robertson and Dowling, 2003). An eight-segment model was created from the unloaded static trial. Trunk mass was modified in the loaded conditions to account for the additional mass. We used the CODA model to define the pelvis segment and hip joint centers (Bell et al., 1989). Hip and knee angles were determined with respect to the proximal segment using a Visual3D hybrid model with a Cardan X-Y-Z (mediolateral, antero-posterior, vertical) rotation sequence (Cole et al., 1993). Internal hip and knee moments were derived using inverse dynamics analysis of kinematics, mass, inertial properties (Dempster et al., 1959) and GRF data. Hip and knee angles and moments in the sagittal and frontal planes were normalized to stride and exported.

A custom MATLAB program (v2017, The Mathworks Inc. Natick, MA) was used to extract sagittal and frontal plane hip and knee angle excursions, peak angles during stance and/or across the gait cycle and peak moments during stance. Moments were normalized two ways: to the participant's body mass in the unloaded condition and to the total mass for each condition (Ogamba et al., 2016). Data from the dominant limb (McLean et al., 2007) were analyzed.

2.6. Statistical analysis

To evaluate the effect of sex while controlling for body mass, the statistical approach was different for non-normalized variables (angle excursion/peaks) and normalized variables (peak moments). For the non-normalized variables, we used a 2x3 mixed design analysis of co-variance (ANCOVA) with sex as a between-subject factor, load as a within-subject factor, and unloaded body mass as a covariate to test for the main effect of sex. An additional analysis of variance (ANOVA) without body mass as a covariate was used to test for the within-subject effect of load and the interaction of load-by-sex (Schneider et al., 2015). For significant interaction effects detected by the ANOVA, post-hoc testing included: a repeated measures ANOVA for each sex with load as the within-subject factor and a univariate ANCOVA for each load with sex as a between-subject factor and unloaded body mass as a covariate. Least significant difference (LSD) pairwise comparisons were used to detect differences between loads if the main effect for load was significant. For normalized variables, a 2x3 mixed design ANOVA with sex as a between-subject factor and load as a within-subject factor was used. For significant load-by-sex interactions, post-hoc testing included a repeated measures ANOVA for each sex with load as the within-subject factor and an independent t-test for each load with sex as the group variable. LSD pairwise comparisons were used to detect differences between loads if the main effect for load was significant. Partial-eta squared (partial-η^2) effects sizes are reported for significant interaction and main effects. Small, medium and large effects were defined as 0.0099, 0.0588, and 0.1379 (Richardson, 2011).

For demographic data, paired t-tests were used to compare body mass, body composition, and body mass index (BMI), and Mann-Whitney U non-parametric tests to compare activity level and RPE. All statistical analyses were performed in SPSS 24.0 (IBM Corp., Chicago, IL) using an *a priori* alpha level of 0.05.

3. Results

Females were significantly lighter, shorter, and had higher body fat percentages than males (Table 1). There were no differences between sexes in age, BMI, activity level, or maximum RPE for any condition. Eighty percent of females and males satisfied both inclusion criteria (load carriage and strenuous exercise). Of the 12 females and 12 males reporting load carriage experience, the highest load carried at least three times was between 9.09 and 15 kg for 2 males, between 15 and 27 kg for 8 females and 3 males, and greater than 27 kg for 4 females and 7 males. The medium and heavy load conditions were on average 19.3% and 36.8% of body mass for females and 16.8% and 30.6% for males (Table 1). The increase in load magnitude between the conditions was 17.6% and 13.8% of body mass for females and males, respectively.

Significant interactions of load and sex were detected in the frontal plane at the hip but in the sagittal and frontal planes at the knee (Tables 2 and 3). While kinetic alterations to load were only noted in females at the hip (Table 3), kinematic changes were noted in females at the knee and in males at the hip (Table 2). Load affected peak hip abduction moment normalized to total mass for females ($p < 0.001$, partial-η^2 effect size (ES) = 0.184), but not for males. In females, peak hip abduction moment normalized to total mass was less in both load conditions compared to unloaded ($p < 0.001$), and less in the heavy load condition compared to medium load ($p = 0.002$). Frontal plane hip excursion ($p = 0.002$, ES = 0.457) and peak hip adduction angle ($p = 0.003$, ES = 0.345) were greater with load in males, but not females. In males, frontal plane hip excursion and peak hip adduction angle were greater in both loaded conditions compared to unloaded ($p \leq 0.03$); only hip excursion was greater in the heavy load compared to medium load ($p = 0.004$). At the knee, load affected sagittal plane excursion only in females ($p < 0.001$, ES = 0.061). Sagittal plane knee excursion was greatest in the unloaded condition compared to both load conditions ($p \leq 0.001$), and greater in the medium load compared to the heavy load ($p = 0.002$).

When comparing females and males walking with the same load, differences were only detected for peak hip adduction angle. In the unloaded ($p = 0.015$) and medium load ($p = 0.036$) conditions females had 3.4° and 3.0° more peak hip adduction than males, respectively.

Significant main effects of load were detected for both kinematic and kinetic variables (Tables 2 and 3). Load affected sagittal plane hip excursion ($p < 0.001$, ES = 0.785) and peak hip flexion angles ($p < 0.001$, ES = 0.423). Hip excursion and peak hip flexion angle were greater in both loaded conditions compared to unloaded ($p < 0.001$), and greater in the heavy load compared to medium load ($p \leq 0.003$). Load affected both peak hip flexion moments normalized to body mass (BM) and to total mass (TM) (BM: $p < 0.001$, ES = 0.832; TM: $p < 0.001$, ES = 0.437) and peak hip extension moments (BM: $p < 0.001$, ES = 0.811; TM: $p < 0.001$, ES = 0.752). Load also affected peak hip abduction moment normalized to body mass ($p < 0.001$, ES = 0.926) and peak hip adduction moment normalized to total mass ($p < 0.001$, ES = 0.427). For each main effect, moments were significantly different between all conditions regardless of normalization. When normalized to body mass, moments were greater with each addition of load ($p < 0.001$). When normalized to total mass, moments were less with each addition of load ($p \leq 0.025$).

In the sagittal plane, load affected peak knee extension angle ($p < 0.001$, ES = 0.560) and peak knee flexion angle across the gait cycle ($p = 0.003$, ES = 0.517) and during early stance ($p = 0.004$, ES = 0.423). Peak knee extension angle was greatest in the unloaded condition compared to both load conditions, and greater in the medium load compared to heavy load ($p < 0.001$). Peak knee flexion angle across the gait cycle and during early stance was greatest in the heavy load condition compared to the medium ($p \leq 0.026$) and unloaded conditions ($p \leq 0.005$).

In the frontal plane, load affected knee excursion ($p = 0.005$, ES = 0.0.196) and peak knee abduction angle ($p < 0.001$, ES = 0.305). Knee excursion ($p \leq 0.006$) and peak knee abduction angle ($p \leq 0.003$) were greatest in the heavy load condition compared to the medium and unloaded conditions. Peak knee abduction angle was greater in the medium load compared to unloaded ($p = 0.025$). Load affected all peak knee moments normalized to body mass ($p \leq 0.016$, ES $\geq$ 0.185), but only knee adduction moment ($p = 0.042$, ES = 0.134) when normalized to total mass. For all peak knee moments normalized to body mass, moments were greater with each addition of load ($p \leq 0.031$). Peak knee adduction moment normalized to total mass was smallest in the heavy load compared to the medium ($p < 0.001$) and unloaded conditions ($p = 0.027$).

4. Discussion

The purpose of this study was to determine if females and males use different hip and knee mechanics when walking with military-relevant symmetric loads. We found that females and males differed in how load affected frontal plane hip mechanics and sagittal plane knee kinematics. This study highlights three important considerations when investigating load carriage: (1) effects of load in the frontal plane, (2) analysis of sex vs. body mass, and (3) joint moment normalization.

First, previous load carriage research has largely focused on sagittal plane effects of load. We analyzed the frontal plane and found not only that load affected frontal plane mechanics but also highlighted sex differences which may have implications for injury mechanisms in female Soldiers. These sex-specific differences in load adaptation also support the assertion that females and males use different gait control strategies (Bruening et al., 2015; Kerrigan et al., 1998). Second, to investigate differences between sexes, we had to account for differences in body mass. While some studies have used body mass to determine the added load, Soldiers carry loads based on the mission, not their mass. Using standardized loads therefore improved our relevance to military applications, but required us to account for body mass. We included body mass as a covariate in our kinematic analysis so that any differences we found due to sex were not the result of differences in body mass. The same reasoning was used when deciding to normalize joint moments to body mass and total mass. Third, kinetic normalization methods are an important consideration for load carriage studies (Liew et al., 2016). We found that load affected hip and knee moments in the sagittal and frontal planes, but the direction and size of the effect were dependent on how the moments were normalized. Normalizing to body mass reflects the change in magnitude of moments and estimates the muscle force required to counteract the added load. Normalizing to total mass allows us to determine the change relative to the added load and investigate how kinematic changes may reduce the effect of added load

Table 2
Mean ± SD for hip and knee kinematic variables.

Angles (°)	Unloaded (UL)		Medium load (ML)		Heavy load (HL)		Load main effect pair wise comp.	Interaction effects post-hoc		Effect size
	Female (♀)	Male (♂)	Female	Male	Female	Male		Load	Sex	(η^2)
Hip										
Sagittal Exc. (cycle)*	43.7 ± 3.7	42.5 ± 2.3	44.6 ± 3.8	43.7 ± 2.6	46 ± 3.6	44.5 ± 2.6	HL > ML[a] HL > UL[a] ML > UL[a]	–	–	0.785
Flexion (cycle)*	39.2 ± 8.4	33.4 ± 6.1	39.9 ± 8.0	34.3 ± 6.0	41.2 ± 7.8	34.7 ± 5.9	HL > ML[b] HL > UL[a] ML > UL[a]	–	–	0.423
Extension (stance)	4.5 ± 8.1	9.1 ± 6.6	4.8 ± 8.0	9.4 ± 6.4	4.8 ± 8.0	9.8 ± 6.0	–	–	–	
Frontal Exc. (cycle)‡	12.8 ± 2.1	11.9 ± 2.2	12.5 ± 2.6	12.5 ± 2.6	12.6 ± 2.7	13.0 ± 2.6	–	♂: HL > ML[b] ♂: HL > UL[b] ♂: ML > UL[b]	–	0.263
Adduction (stance)‡	7.1 ± 2.7	3.7 ± 2.6	7.1 ± 2.6	4.1 ± 2.7	6.9 ± 2.8	4.3 ± 2.8	–	♂: HL > UL[b] ♂: ML > UL[b]	UL: ♀>♂[b] ML: ♀>♂[b]	0.167
Abduction (cycle)	5.7 ± 2.7	8.2 ± 3.6	5.4 ± 2.8	8.4 ± 3.6	5.7 ± 2.9	8.7 ± 3.7	–	–	–	–
Knee										
Sagittal Exc. (cycle)	68.97 ± 2.67	69.81 ± 3.81	68.02 ± 2.56	69.93 ± 3.46	67.29 ± 2.64	69.49 ± 3.48	–	♀: HL < ML[b] ♀: HL < NL[a] ♀: ML < NL[b]	–	0.145
Extension (cycle)*	0.53 ± 3.54	1.69 ± 4.04	−0.36 ± 3.06	1.02 ± 4.11	−1.96 ± 3.32	0.19 ± 4.12	HL < ML[a] HL < NL[a] ML < NL[a]	–	–	0.560
Flexion (cycle)*	68.44 ± 3.13	68.12 ± 2.51	68.38 ± 3.39	68.9 ± 2.53	69.25 ± 2.83	69.3 ± 2.56	HL > ML[b] HL > NL[b]	–	–	0.217
Flexion (Early Stance)*	19.41 ± 3.94	18.71 ± 4.56	19.42 ± 3.88	19.16 ± 4.61	20.73 ± 3.91	19.33 ± 4.67	HL > ML[b] HL > NL[b]	–	–	0.209
Frontal Exc. (cycle)*	16.67 ± 6.35	17.87 ± 5.23	17.01 ± 6.5	18.14 ± 5.18	17.85 ± 6.64	18.53 ± 4.96	HL > ML[b] HL > NL[b]	–	–	0.196
Adduction (Stance)‡	−1.95 ± 1.68	0.29 ± 3.31	−1.94 ± 1.63	0.31 ± 3.25	−2.18 ± 1.79	0.55 ± 3.17		No Sig. Post-hoc		0.116
Abduction (cycle)*	17.51 ± 7.26	15.68 ± 5.1	17.98 ± 7.53	16.24 ± 5.17	19.21 ± 7.91	16.56 ± 4.96	HL > ML[b] HL > NL[a] ML > NL[b]	–		0.305

Note:
Abbreviations: Exc. = Excursion, Sig. = Significant.
* Indicates significant ($p < 0.05$) main effect for load.
‡ Indicates significant ($p < 0.05$) load-by-sex interaction effect.
[a] Indicates $p < 0.001$.
[b] Indicates $p < 0.05$.

Table 3
Mean ± SD for hip and knee kinetic variables.

Moments (Nm/kg)	Unloaded (UL) Female (♀)	Unloaded (UL) Male (♂)	Medium load (ML) Female	Medium load (ML) Male	Heavy load (HL) Female	Heavy load (HL) Male	Load main effect pair wise comp.	Interaction effects post-hoc: Load	Interaction effects post-hoc: Sex	Effect Size (η^2)
Hip										
Normalized to BM										
Flexion*	0.67 ± 0.17	0.71 ± 0.16	0.76 ± 0.19	0.79 ± 0.17	0.83 ± 0.20	0.87 ± 0.17	HL > ML[a] HL > UL[a] ML > UL[a]	–	–	0.832
Extension*	0.70 ± 0.11	0.76 ± 0.14	0.77 ± 0.12	0.81 ± 0.14	0.85 ± 0.14	0.87 ± 0.15	HL > ML[a] HL > UL[a] ML > UL[a]	–	–	0.811
Adduction	0.09 ± 0.03	0.11 ± 0.04	0.10 ± 0.05	0.12 ± 0.05	0.10 ± 0.06	0.12 ± 0.06	–	–	–	
Abduction*	0.92 ± 0.11	0.83 ± 0.14	1.05 ± 0.13	0.96 ± 0.14	1.17 ± 0.16	1.06 ± 0.16	HL > ML[a] HL > UL[a] ML > UL[a]	–	–	0.926
Normalized to TM										
Flexion*	0.67 ± 0.17	0.71 ± 0.16	0.64 ± 0.17	0.68 ± 0.15	0.61 ± 0.15	0.67 ± 0.14	HL < ML[b] HL < UL[a] ML < UL[a]	–	–	0.437
Extension*	0.70 ± 0.11	0.76 ± 0.14	0.65 ± 0.10	0.70 ± 0.13	0.62 ± 0.10	0.66 ± 0.12	HL < ML[a] HL < UL[a] ML < UL[a]	–	–	0.752
Adduction*	0.09 ± 0.03	0.11 ± 0.04	0.08 ± 0.04	0.10 ± 0.04	0.07 ± 0.04	0.09 ± 0.04	HL < ML[a] HL < UL[a] ML < UL[b]	–	–	0.427
Abduction‡	0.92 ± 0.11	0.83 ± 0.14	0.88 ± 0.11	0.82 ± 0.12	0.86 ± 0.12	0.81 ± 0.12	–	♀: HL < ML[a] ♀: HL < UL[b] ♀: ML < UL[b]	–	0.184
Knee										
Normalized to BM										
Flexion*	0.57 ± 0.14	0.66 ± 0.22	0.66 ± 0.17	0.77 ± 0.25	0.8 ± 0.19	0.87 ± 0.27	HL > ML[a] HL > NL[a] ML > NL[a]	–	–	0.833
Extension*	0.4 ± 0.1	0.44 ± 0.1	0.49 ± 0.13	0.51 ± 0.14	0.56 ± 0.16	0.56 ± 0.17	HL > ML[a] HL > NL[a] ML > NL[a]	–	–	0.761
Adduction*	0.1 ± 0.02	0.1 ± 0.05	0.11 ± 0.05	0.11 ± 0.05	0.12 ± 0.05	0.11 ± 0.06	HL > ML HL > NL ML > NL	–	–	0.185
Abduction*	0.37 ± 0.1	0.38 ± 0.11	0.44 ± 0.1	0.44 ± 0.13	0.51 ± 0.11	0.51 ± 0.15	HL > ML[a] HL > NL[a] ML > NL[a]	–	–	0.841
Normalized to TM										
Flexion	0.57 ± 0.14	0.66 ± 0.22	0.55 ± 0.14	0.66 ± 0.22	0.59 ± 0.14	0.67 ± 0.21	–	–	–	–
Extension	0.4 ± 0.1	0.44 ± 0.1	0.41 ± 0.11	0.44 ± 0.12	0.41 ± 0.12	0.43 ± 0.13	–	–	–	–
Adduction*	0.1 ± 0.02	0.1 ± 0.05	0.09 ± 0.04	0.09 ± 0.05	0.09 ± 0.04	0.09 ± 0.04	HL < ML[a] HL < NL[b]	–	–	0.134
Abduction	0.37 ± 0.1	0.38 ± 0.11	0.37 ± 0.08	0.38 ± 0.11	0.37 ± 0.08	0.39 ± 0.11	–	–	–	–

Note:
Abbreviations: BM = Body Mass, TM = Total Mass.
* Indicates significant ($p < 0.05$) main effect for load.
‡ Indicates significant ($p < 0.05$) load-by-sex interaction effect.
[a] Indicates $p < 0.001$.
[b] Indicates $p < 0.05$.

(Ogamba et al., 2016). While all significant results except one had large effect sizes, the effects were greater when normalized to body mass than to total mass (Table 3) suggesting that kinematic changes reduced effect of the added load.

We found that the peak hip abduction moment normalized to body mass increased in both females and males with increasing load; however, when normalized to total mass, the peak hip abduction moment decreased in females and did not change in males. During stance, the hip abduction moment requires hip abductor muscle activity to control the drop of the pelvis (Seay, 2015) due to the superincumbent weight of the trunk, head, arms and swing leg. While females and males have different pelvic structures (Lewis et al., 2017) which could affect the hip abductor muscles, a dynamic biomechanical model did not find differences in locomotor efficiency (Warrener et al., 2015). Our finding suggests that females modify their mechanics elsewhere to reduce the effect of the increased load on the hip abduction moment. For example, increased ipsilateral trunk flexion (i.e., compensated Trendelenburg) (Neumann, 2013) may be used to reduce the hip abduction moment. While hip kinematics in females did not change, the peak hip adduction angle and frontal plane excursion increased with increased load in males. With increasing load, the peak hip adduction angle of males became similar to females with no sex difference detected at the heaviest load. This lack of change with load may suggest females are at end range of hip adduction during unloaded walking. During dynamic tasks, increased peak hip adduction angle is associated with increased risk of PFP (Powers, 2010). The higher peak hip adduction angle seen in females for all loads may increase their risk of PFP.

In both males and females, the peak hip adduction moment decreased with increasing load when normalized to total mass, but did not change when normalized to body mass. The decrease in the hip adduction moment during early stance may indicate kinematic or muscular changes to reduce the effect of load. While these moments are small in magnitude (≤0.12 Nm/kg), the effect may accumulate over prolonged load carriage exercise. Alterations in hip adductor muscle forces during early stance may increase the risk of pelvic stress fractures in females (Kelly et al., 2000).

Similar to previous studies, we found that load affected sagittal plane hip kinematics and kinetics. Consistent with our results, peak hip flexion angle (Attwells et al., 2006; Harman et al., 2000; Silder et al., 2013), and both peak hip extension (Silder et al., 2013) and peak hip flexion moments (Harman et al., 2000; Seay et al., 2014) normalized to body mass increase with increased load. Studies of backpack loads attribute the increased hip flexion angle to forward trunk lean from the posterior load (Attwells et al., 2006; Harman et al., 2000). The increased hip extension moment controls the descent of the load during weight acceptance (Seay, 2015). While increases in the peak hip flexion angle and moment have been noted with symmetric loads ≥30% of body mass (Seay et al., 2014; Silder et al., 2013), we detected large differences at the medium load (~18% of body mass) compared to the unloaded condition. While we noted differences with lighter loads than previously used, our larger sample size contained a wide range of load magnitude as a percentage of body mass. The large effect sizes of our results is likely related to the use of military-relevant standardized loads as opposed to loads relative to body mass and large sample size.

At the knee, we found that load had a large effect on peak knee abduction and adduction moments. Both moments increased with load when normalized to body mass, while the peak knee adduction moment normalized to total mass decreased with load. The increased peak knee abduction moment normalized to body mass during late stance may increase compressive force on the medial knee compartment (Schipplein and Andriacchi, 1991) and increase the risk of knee osteoarthritis (Chang et al., 2004). The knee adduction moment normalized to body mass noted in early stance resists valgus collapse and can contribute to anterior cruciate or medial collateral ligament injury (Powers, 2010). Similar to the hip, the decreased knee adduction moment normalized to total mass may indicate kinematic (i.e. knee flexion angle) or muscular changes which reduce the effect of load. While the magnitude of the knee adduction moments are small (≤0.12 Nm/kg), the large effect size of this result suggests these changes are robust.

Similar to previous studies, we found that knee flexion angle in early stance (Kinoshita, 1985; Silder et al., 2013) and peak knee extension moment normalized to body mass (Seay et al., 2014; Silder et al., 2013) increased with increased load. Increased knee flexion angle aids in shock absorption (Attwells et al., 2006; Kinoshita, 1985) and keeps the COM closer to the ground (Harman et al., 2000). However, the increased knee flexion angle requires greater quadriceps activation, increasing metabolic cost (Silder et al., 2013). The increased knee extension moment accompanies the increased hip extension moment to control the decent of added load. (Seay, 2015)

We did note one sex-specific difference in knee kinematics. In females, sagittal plane knee excursion decreased with load. Knee excursion is typically unaffected by load despite an increase in the knee extension moment during early stance (Seay, 2015), which is thought to increase stress on knee structures (Seay et al., 2014). While both sexes increased the knee extension moment during early stance and decreased the peak knee extension angle just prior to heel strike, only females decreased knee excursion. It is possible that in females the decreased excursion is coming from the decreased extension angle just prior to heel strike and would further increase stress on knee structures. The difference in frontal plane hip and sagittal plane knee load adaptation strategies between sexes may contribute to the higher prevalence of PFP seen in female Soldiers.

The generalizability of this study is limited due to the loads used and the testing duration. While the 15 and 26 kg loads elicited sex-specific modifications, Soldiers often wear loads exceeding 50 kg (Dean and DuPont, 2004). The use of symmetric loads may not generalize to Soldiers carrying backpack loads; however, symmetric loads are becoming more widely used (Hasselquist et al., 2018; Seay et al., 2014). Despite including more strides than most studies, the effects are from a 2-min walking trial and do not account for long-term adaptations with prolonged load carriage. Extensive load carriage experience may affect how an individual adapts to the load. Given that all our participants were physically active, had load carriage experience, or both, we do not anticipate that our results were substantially impacted by lack of physical fitness or load carriage experience.

In summary, this is the first study to detect sex-specific differences in hip and knee mechanics while walking with military-relevant loads. These results indicate that females and males change frontal plane hip and sagittal plane knee mechanics differently while walking with a load. While kinetic alterations were noted in females at the hip, kinematic changes were noted in males at the hip and in females at the knee. The differences in these strategies may increase the risk of hip or knee injury in females compared to males. We noted load effects that were previously undetected, highlighting the importance of using military-relevant standardized loads and investigating frontal plane adaptations. This study sets a foundation for future research investigating how females respond to heavier, military-relevant loads.

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Declaration of Competing Interest

The authors have no conflicts to disclose.

References

Attwells, R.L., Birrell, S.A., Hooper, R.H., Mansfield, N.J., 2006. Influence of carrying heavy loads on soldiers' posture, movements and gait. Ergonomics 49, 1527–1537. https://doi.org/10.1080/00140130600757237.

Bell, A.L., Brand, R.A., Pedersen, D.R., 1989. Prediction of hip joint centre location from external landmarks. Hum. Mov. Sci. 8, 3–16.

Birrell, S.A., Haslam, R.A., 2009. The effect of military load carriage on 3-D lower limb kinematics and spatiotemporal parameters. Ergonomics 52, 1298–1304. https://doi.org/10.1080/00140130903003115.

Birrell, S.A., Haslam, R.A., 2010. The effect of load distribution within military load carriage systems on the kinetics of human gait. Appl. Ergon. 41, 585–590. https://doi.org/10.1016/j.apergo.2009.12.004.

Boling, M., Padua, D., Marshall, S., Guskiewicz, K., Pyne, S., Beutler, A., 2010. Gender differences in the incidence and prevalence of patellofemoral pain syndrome. Scand. J. Med. Sci. Sport. 20, 725–730. https://doi.org/10.1111/j.1600-0838.2009.00996.x.

Borg, G.A., 1982. Psychophysical bases of perceived exertion. Med. Sci. Sports Exerc. 14, 377–381.

Bruening, D.A., Frimenko, R.E., Goodyear, C.D., Bowden, D.R., Fullenkamp, A.M., 2015. Sex differences in whole body gait kinematics at preferred speeds. Gait Posture 41, 540–545. https://doi.org/10.1016/j.gaitpost.2014.12.011.

Chang, A., Hayes, K., Dunlop, D., Hurwitz, D., Song, J., Cahue, S., Genge, R., Sharma, L., 2004. Thrust during ambulation and the progression of knee osteoarthritis. Arthritis Rheum. 50, 3897–3903. https://doi.org/10.1002/art.20657.

Cole, G.K., Nigg, B.M., Ronsky, J.L., Yeadon, M.R., 1993. Application of the joint coordinate system to three-dimensional joint attitude and movement

representation: a standardization proposal. J. Biomech. Eng. 115, 344. https://doi.org/10.1115/1.2895496.
Dean, C., DuPont, F., 2004. The modern warrior's combat load: Dismounted Operations in Afghanistan, Technical Report. Fort Levenworth, Kansas.
Dempster, W.T., Gabel, W.C., Felts, W.J.L., 1959. The anthropometry of the manual work space for the seated subject. Am. J. Phys. Anthropol. 17, 289–317. https://doi.org/10.1002/ajpa.1330170405.
Department of the Army, 1990. Foot marches (Field Manual 21-18), US Government Printing Office.
Edwards, W.B., Gillette, J.C., Thomas, J.M., Derrick, T.R., 2008. Internal femoral forces and moments during running: Implications for stress fracture development. Clin. Biomech. 23, 1269–1278. https://doi.org/10.1016/j.clinbiomech.2008.06.011.
Harman, E., Han, K.H., Frykman, P., Pandorf, C., 2000. The effects of backpack weight on the biomechanics of load carriage. DTIC Doc. T00–17, 72.
Hasselquist, L., Eddy, M., Mitchell, B., Brown, S., McNamara, J., Hancock, C., Caruso, C., 2018. Assessing the impact of clothing and individual equipment on soldier physical, biomechanical, and congnitive performance part I: test methodology. DTIC Doc., TR-18/004
Hill, P.F., Chatterji, S., Chambers, D., Keeling, J.D., 1996. Stress fracture of the pubic ramus in female recruits. J. Bone Joint Surg. Br. 78, 383–386.
Jackson, A.S., Pollock, M.L., 1985. Practical assessment of body composition. Phys. Sportsmed. 13, 76–90. https://doi.org/10.1080/00913847.1985.11708790.
Jones, B.H., Hauret, K.G., Dye, S.K., Hauschild, V.D., Rossi, S.P., Richardson, M.D., Friedl, K.E., 2017. Impact of physical fitness and body composition on injury risk among active young adults: A study of Army trainees. J. Sci. Med. Sport 20, S17–S22. https://doi.org/10.1016/j.jsams.2017.09.015.
Kelly, E.W., Jonson, S.R., Cohen, M.E., Shaffer, R., 2000. Stress fractures of the pelvis in female navy recruits: an analysis of possible mechanisms of injury. Mil. Med. 165, 142–146.
Kerrigan, D.C., Todd, M.K., Della Croce, U., 1998. Gender differences in joint biomechanics during walking: Normative study in young adults. Am. J. Phys. Med. Rehabil. https://doi.org/10.1097/00002060-199801000-00002.
Kinoshita, H., 1985. Effects of different loads and carrying systems on selected biomechanical parameters describing walking gait. Ergonomics 28, 1347–1362. https://doi.org/10.1080/00140138508963251.
Knapik, J.J., Reynolds, K., 2015. Load carriage-related injury mechanisms, risk factors, and prevention. In: The Mechanobiology and Mechanophysiology of Military-Related Injuries. Springer, pp. 107–137. https://doi.org/10.1007/8415_2014_182.
Krupenevich, R., Rider, P., Domire, Z., DeVita, P., 2015. Males and females respond similarly to walking with a standardized, heavy load. Mil. Med. 180, 994–1000. https://doi.org/10.7205/MILMED-D-14-00499.
Kupferer, K.R., Bush, D.M., Cornell, J.E., Lawrence, V.A., Alexander, J.L., Ramos, R.G., Curtis, D., 2014. Femoral neck stress fracture in Air Force basic trainees. Mil. Med. 179, 56–61. https://doi.org/10.7205/MILMED-D-13-00154.
LaFiandra, M., Wagenaar, R., Holt, K., Obusek, J., 2003. How do load carriage and walking speed influence trunk coordination and stride parameters? J. Biomech. 36, 87–95. https://doi.org/10.1016/S0021-9290(02)00243-9.
Lee, C.-H., Huang, G.-S., Chao, K.-H., Jean, J.-L., Wu, S.-S., 2003. Surgical treatment of displaced stress fractures of the femoral neck in military recruits: a report of 42 cases. Arch. Orthop. Trauma Surg. 123, 527–533. https://doi.org/10.1007/s00402-003-0579-8.
Lenton, G.K., Doyle, T.L.A., Lloyd, D.G., Higgs, J., Billing, D., Saxby, D.J., 2019. Lower-limb joint work and power are modulated during load carriage based on load configuration and walking speed. J. Biomech. 83, 174–180. https://doi.org/10.1016/j.jbiomech.2018.11.036.
Lewis, C.L., Laudicina, N.M., Khuu, A., Loverro, K.L., 2017. The human pelvis: variation in structure and function during gait. Anat. Rec. 300, 633–642. https://doi.org/10.1002/ar.23552.
Liew, B., Morris, S., Netto, K., 2016. The effect of backpack carriage on the biomechanics of walking: A systematic review and preliminary meta-analysis. J. Appl. Biomech. 32, 614–629. https://doi.org/10.1123/jab.2015-0339.
Loverro, K.L., Brown, T.N., Coyne, M.E., Schiffman, J.M., 2015. Use of body armor protection with fighting load impacts soldier performance and kinematics. Appl. Ergon. 46, 168–175. https://doi.org/10.1016/j.apergo.2014.07.015.
Majumdar, Deepti, Pal, M.S., Majumdar, Dhurjati, 2010. Effects of military load carriage on kinematics of gait. Ergonomics 53, 782–791. https://doi.org/10.1080/00140131003672015.
McLean, S.G., Fellin, R.E., Felin, R.E., Suedekum, N., Calabrese, G., Passerallo, A., Joy, S., 2007. Impact of fatigue on gender-based high-risk landing strategies. Med. Sci. Sports Exerc. 39, 502–514. https://doi.org/10.1249/mss.0b013e3180d47f0.
Neumann, D.A., 2013. Kinesiology of the musculoskeletal system: foundations for rehabilitation. Elsevier Health Sciences.
Nindl, B.C., Jones, B.H., Van Arsdale, S.J., Kelly, K., Kraemer, W.J., 2016. Operational physical performance and fitness in military women: physiological, musculoskeletal injury, and optimized physical training considerations for successfully integrating women into combat-centric military occupations. Mil. Med. 181, 50–62. https://doi.org/10.7205/MILMED-D-15-00382.
Ogamba, M.I., Loverro, K.L., Laudicina, N.M., Gill, S.V., Lewis, C.L., 2016. Changes in gait with anteriorly added mass: a pregnancy simulation study. J. Appl. Biomech. 32, 379–387. https://doi.org/10.1123/jab.2015-0178.
Powers, C.M., 2010. The influence of abnormal hip mechanics on knee injury: a biomechanical perspective. J. Orthop. Sport. Phys. Ther. 40, 42–51. https://doi.org/10.2519/jospt.2010.3337.
Richardson, J.T.E., 2011. Eta squared and partial eta squared as measures of effect size in educational research. Educ. Res. Rev. 6, 135–147. https://doi.org/10.1016/j.edurev.2010.12.001.
Robertson, D.G.E., Dowling, J.J., 2003. Design and responses of Butterworth and critically damped digital filters. J. Electromyogr. Kinesiol. 13, 569–573. https://doi.org/10.1016/S1050-6411(03)00080-4.
Schipplein, O.D., Andriacchi, T.P., 1991. Interaction between active and passive knee stabilizers during level walking. J. Orthop. Res. 9, 113–119. https://doi.org/10.1002/jor.1100090114.
Schneider, B.A., Avivi-Reich, M., Mozuraitis, M., 2015. A cautionary note on the use of the Analysis of Covariance (ANCOVA) in classification designs with and without within-subject factors. Front. Psychol. 6, 1–12. https://doi.org/10.3389/fpsyg.2015.00474.
Schuh-Renner, A., Grier, T.L., Canham-Chervak, M., Hauschild, V.D., Roy, T.C., Fletcher, J., Jones, B.H., 2017. Risk factors for injury associated with low, moderate, and high mileage road marching in a U.S. Army infantry brigade. J. Sci. Med. Sport 20, S28–S33. https://doi.org/10.1016/j.jsams.2017.07.027.
Scott, S.J., Feltwell, D.N., Knapik, J.J., Barkley, C.B., Hauret, K.G., Bullock, S.H., Evans, R.K., 2012. A multiple intervention strategy for reducing femoral neck stress injuries and other serious overuse injuries in US Army Basic Combat Training. Mil. Med. 177, 1081–1089.
Seay, J.F., 2015. Biomechanics of load carriage. In: Studies in Mechanobiology, Tissue Engineering and Biomaterials, The Mechanobiology and Mechanophysiology of Military-Related Injuries. Springer, pp. 1–36.
Seay, J.F., Fellin, R.E., Sauer, S.G., Frykman, P.N., Bensel, C.K., 2014. Lower extremity biomechanical changes associated with symmetrical torso loading during simulated marching. Mil. Med. 179, 85–91. https://doi.org/10.7205/MILMED-D-13-00090.
Silder, A., Delp, S.L., Besier, T., 2013. Men and women adopt similar walking mechanics and muscle activation patterns during load carriage. J. Biomech. 46, 2522–2528. https://doi.org/10.1016/j.jbiomech.2013.06.020.
Terwee, C.B., Bouwmeester, W., Van Elsland, S.L., De Vet, H.C.W., Dekker, J., 2011. Instruments to assess physical activity in patients with osteoarthritis of the hip or knee: a systematic review of measurement properties. Osteoarthr. Cartil. 19, 620–633. https://doi.org/10.1016/j.joca.2011.01.002.
Warrener, A.G., Lewton, K.L., Pontzer, H., Lieberman, D.E., 2015. A wider pelvis does not increase locomotor cost in humans, with implications for the evolution of childbirth. PLoS One 10, 1–14. https://doi.org/10.1371/journal.pone.0118903.