MILITARY MEDICINE, 180, 9:994, 2015

Males and Females Respond Similarly to Walking With a Standardized, Heavy Load

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ABSTRACT Females in the military sustain a higher incidence of lower extremity injuries compared to males. Previous investigations of gender differences during load carriage used loads normalized to body mass; as a result of anthropometric and strength differences between genders, this may partially normalize to strength, masking gender or size differences in response to load. We compared gait kinetics and kinematics between genders based on a standardized load, instead of loads relative to body mass. 11 males and 11 females walked at 1.5 m/s over level ground with a 22 kg rucksack using three load conditions: unloaded, low-back placement, and mid-back placement. We found a gender by load interaction for average trunk position ($p < 0.05$). Stride length decreased 1.3% in loaded vs. unloaded walking. Loaded walking increased knee extensor (65%) and ankle plantarflexor torque (23%, all $p < 0.0001$), but not hip extensor torque ($p > 0.05$) compared to unloaded walking. The lack of gender differences may indicate that females do not adapt gait mechanics to account for smaller stature and lesser absolute strength compared to males, which may contribute to the high injury rate in female military recruits.

INTRODUCTION

Load carriage, or walking while carrying additional loads, is often associated with recreational pursuits, such as hiking, or professions that require transporting large amounts of equipment, for example, firefighting or military service. Individuals in fields such as emergency medical care, factory work, and those in the military are frequently required to carry loads ranging from 5 to 35 kg as part of their daily job duties.[1–3] Often, load amounts in these professions are based on the position or job being performed rather than on individual anthropometric or physical capacity measures. Specifically, individuals in the military carry increasingly heavy loads and can even be required to carry loads up to 68 kg regardless of their mass or muscle strength.[2,4] Carrying these heavy loads places a large amount of stress on the trunk and lower extremities. Ground reaction forces (GRFs)[5,6] and forces at the lumbosacral spine[7] increase proportionally as the load being carried increases as do hip, knee, and ankle joint torques.[8] Consequently, load carriage in the military results in high rates of lower extremity injury.[9–11]

It has been well documented that female soldiers incur higher rates of injury than their male counterparts.[12–14] In fact, being a female is one of the top risk factors for sustaining injuries in the military[10,15] as females are twice as likely to sustain an injury and almost 2.5 times as likely to sustain a more serious time-loss injury compared to males.[12] Heavy rucksack loads and soldiers on foot patrol, an activity requiring soldiers to carry large amounts of equipment, account for approximately 20% of documented injuries[9,10] Of these injuries, females typically sustain a high incidence of lower extremity stress fractures with injuries to the pelvis, metatarsals, and tibia being most prominent.[16] Females also experience a large incidence of nonstress fracture overuse injuries at the knee and shank compared to males.[10,17] The large discrepancy in injury rates between males and females suggests the standardized loads may overload female soldiers relative to their capacity.

Although there have been several investigations of the effect of carrying a heavy load on gait biomechanics in males[5,8,18–20] or the effect of load carriage on gait biomechanics in females,[21,22] there have been less that focus on gender differences while walking with a load relative to body mass,[23,24] and none that have compared gait kinematics, kinetics, and energetics between males and females walking while carrying an identical, standardized load on the back. Silder et al[23] reported no gender differences in stride length or rate, joint angular positions, joint moments, or muscle electromyographic data when males and females carried loads at 10%, 20%, and 30% of their body mass. These findings, however, may differ from those in which all participants carried the same standardized load as suggested by Silder et al.[23] Body mass is correlated with strength[25] and military performance[26]; therefore, body mass–normalized data are most likely also normalized to muscle strength or physical capacity. This is problematic given that in the general population females have less body mass, are on average weaker,[27,28] and are shorter in height compared to males.[12,25] Several characteristics of smaller stature individuals, most notably lower muscle strength,[25] may hamper their ability to carry the same load as their larger stature counterparts. This notion is consistent with the observations that regardless of gender, individuals of smaller stature sustain more stress fractures of the lower extremity than larger individuals.[29,30] We now speculate that when carrying a heavy load females experience more load per unit muscle strength than males, resulting in a decreased ability to adapt their gait biomechanics to support the load, creating a potential source of injury.

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doi: 10.7205/MILMED-D-14-00499

It is likely that these anthropometric differences, in conjunction with the strength deficits observed in females compared to males, would result in different gait adaptations in males and females when carrying the same standardized load. To attenuate the increases in musculoskeletal loads caused by load carriage, females could be expected to lean forward more than males because of their smaller stature and to shorten their steps more than males because this adaptation reduces external musculoskeletal loads.[31] Additionally, because of their smaller masses females could still walk with larger mass-normalized joint torques and powers than males when carrying an additional load. In fact, Silder et al[23] suggested that using a fixed load amount for all participants while investigating the effects of load carriage on gait biomechanics could identify gender differences. Thus, we hypothesize an interaction effect between gender and load carriage such that females compared to males will respond differently to a standardized, additional heavy load during level walking. The purpose of this study was to compare lower extremity gait biomechanics between males and females walking with and without a standardized load. Specifically, we compared GRFs; trunk position; and hip, knee, and ankle torques; and powers between males and females while walking unloaded and while carrying a 22 kg rucksack over a level walking surface.

METHODS

Participants

A total of 22 healthy adults (11 males, 11 females) participated in this study after providing written informed consent in accordance with University policy. Table I provides subject characteristic data. The majority of our participants (77%) were members of the Army Reserve Officers' Training Corps at East Carolina University and had substantial experience carrying heavy loads, and several of our subjects were recreationally active, healthy, and fit students from the Department of Kinesiology. A medical history questionnaire was given before participation in the study to ensure that all individuals were healthy and free of any neurological diseases or orthopedic injuries to the trunk or lower extremities. Testing for each subject was conducted in one laboratory session lasting approximately 2 hours. Testing protocols were approved by the University Institutional Review Board for human research.

TABLE I. Subject Characteristics

	Age (years)	Mass (kg)	Height (m)	BMI (kg/m^2)
Males (n = 11)	20 ± 2.3	79.1 ± 13.3	1.79 ± 0.09	24.5 ± 2.7
Females (n = 11)	20 ± 1.8	72.9 ± 15.1	1.71 ± 0.08	24.7 ± 4.3

BMI, body mass index. Values are mean ± SD.

Experimental Set-up

For the level walking conditions, three-dimensional GRFs were collected at 960 Hz using a force platform (AMTI, Newton, Massachusetts) embedded in a raised walkway. Kinematic data were captured at 120 Hz using an eight camera infrared digital camera system (Qualisys ProReflex, Gothenburg, Sweden). An infrared timing system (Brower Timing System, Salt Lake City, Utah) was used to constrain walking velocity within 5% of 1.50 m/s throughout all walking tasks. A controlled speed was used to remove the effects of altered walking velocity on load carriage.

Subjects wore T-shirts, spandex bicycle shorts, and their own boots or athletic shoes during testing. Subjects also wore a MOLLE (Modular Lightweight Load Carrying Equipment) rucksack containing a 22 kg mass during loaded conditions. The 22 kg load was chosen to be consistent with the recommended U.S. Army fighting load.[4] To define joint locations, passive reflective markers were placed on the following anatomical landmarks: first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral knee, right and left greater trochanters, and the right and left acromion processes. To track motion of the segments, clusters of four markers were placed on the foot, shank, and thigh, and individual markers were placed on the umbilicus and sternum. Individual tracking markers and the marker cluster on the foot were securely placed with tape on the participant's form-fitting clothing and shoes. Marker clusters on the shank and thigh were securely placed using neoprene sleeves, velcro, and tape. Participants performed a standing calibration trial and then completed at least three walking trials, contacting the force platform with the right foot, in three different load conditions: unloaded, low-back load placement, and mid-back load placement. The low-back load placement concentrated the mass toward the bottom of the pack at approximately the location of the first and second lumbar vertebrae, and the mid-back load placement raised the load in the pack approximately 12 cm from the low-back position using plastic risers inside the pack. The order of the load conditions was counterbalanced among subjects. We report only one load carriage condition in this study, the low-back placement. Foam padding was used in the pack to keep the mass from shifting during testing. Participants completed an average of three practice trials before data collection.

Data Analyses

Data were analyzed using Visual 3D software (C-Motion, Germantown, Maryland). An average of three trials per condition were analyzed for each subject. We first created a biomechanical model using the marker placements from the static calibration trial, anthropometrics,[32] and the subject's height and weight. Before calculating gait biomechanics, the digitized Cartesian coordinates of the reflective markers describing the full gait cycle before and on the force platform (during the stance phase) were processed bidirectionally through a second-order low-pass digital filter with a 6 Hz cutoff frequency. GRF data were processed through a low-pass digital filter with a 45 Hz cutoff frequency. Stride length; walking velocity; trunk angular position; and joint angular positions and angular velocities at the right hip, knee, and ankle were calculated from the linear position data. The Cardan joint rotational sequence used to calculate joint angles was flexion/extension, adduction/abduction, and internal/external rotation.[33] Inverse dynamics within Visual 3D software were then used to calculate the right limb joint reaction forces and internal joint torques at these joints by combining the GRF and linear position and acceleration data with Newtonian equations of motion. Joint powers were calculated as the product of the joint angular velocities and torques. The data were further processed using proprietary laboratory software to identify selected angular positions and maximum GRFs, torques, and powers during the stance phases. Joint torques, GRFs, and joint powers were expressed relative to body mass. Although the carried load was standardized to a single value, we chose to report these variables per unit mass because the females were systematically lighter than the males leading to the expectation that females would be loaded more than males per unit mass.

Statistical Analyses

A two-factor analysis of variance with repeated measures on load was run to test for interactions between two genders and two load conditions and for main effects of gender and load on stride length; average trunk position during stance; mass-normalized maximum GRFs; and hip, knee, and ankle joint torques and joint powers. Pearson Product Moment Correlation analyses were also performed between participants' masses and forward trunk lean, second vertical maximum GRF, maximum ankle plantarflexor torque, and maximum propulsive anterior GRF. Alpha was set at $p < 0.05$ for all analyses.

RESULTS

Interactions Between Gender and Load and Main Effect for Gender

A significant gender by load interaction was detected for average trunk position ($p = 0.025$). Females exhibited an ~13° increase in forward trunk position, whereas males exhibited an ~11° increase in forward trunk position (Table II). We did not observe significant interactions between gender and load for any of the remaining variables (all $p > 0.05$). Figure 1 illustrates the similarity across genders in response to the heavy load for the vertical and anteroposterior body mass-normalized GRFs. Additionally, the lack of interaction between gender and load for hip, knee, and ankle torques normalized to body mass can be observed in Figure 2. None of the kinetic or kinematic variables exhibited a significant difference because of gender.

Main Effect of Load

Stride length decreased 1.3% from the unloaded to loaded condition ($p = 0.037$). There were significant, ~27% increases in the first and second maximums of vertical GRF and in the maximum braking and propulsion forces of anteroposterior GRF in the loaded compared to unloaded condition (all $p < 0.0001$; Fig. 1). Participants used similar torque and power patterns in the unloaded and loaded conditions but these patterns differed in magnitude at the knee and ankle in torque and at all joints in power (Figs. 2 and 3). There were no load effects for hip extensor torque or positive

TABLE II. Gait Kinematics, Kinetics, and Energetics

	Female No Load	Female Loaded	Male No Load	Male Loaded
Kinematics				
Stride Length[a]	1.59 ± 0.07	1.57 ± 0.10	1.62 ± 0.10	1.59 ± 0.09
Trunk, Average Position Stance[a,b]	0.82 ± 2.79	−14.12 ± 3.54	−0.81 ± 3.16	−11.94 ± 4.28
Kinetics				
Vertical GRF, First Maximum[a]	11.75 ± 1.88	14.77 ± 1.05	11.81 ± 0.54	15.12 ± 1.22
Vertical GRF, Second Maximum[a]	11.32 ± 1.24	14.37 ± 1.38	11.20 ± 0.51	14.21 ± 0.82
Ant-Post GRF, Maximum Brake[a]	−2.30 ± 0.42	−3.02 ± 0.35	−2.16 ± 0.20	−3.02 ± 0.35
Ant-Post GRF, Maximum Propel[a]	2.48 ± 0.29	3.06 ± 0.30	2.44 ± 0.25	3.10 ± 0.37
Hip, Maximum Extensor Torque	0.84 ± 0.20	0.88 ± 0.17	0.87 ± 0.12	0.93 ± 0.22
Knee, Maximum Extensor Torque[a]	0.45 ± 0.24	0.72 ± 0.28	0.48 ± 0.23	0.81 ± 0.22
Ankle, Maximum p-Flexor Torque[a]	1.71 ± 0.20	2.12 ± 0.24	1.73 ± 0.18	2.13 ± 0.17
Energetics				
Hip, Maximum Positive Power 1	0.91 ± 0.39	0.88 ± 0.34	0.91 ± 0.37	0.89 ± 0.61
Hip, Maximum Positive Power 2[a]	1.11 ± 0.37	1.28 ± 0.36	1.08 ± 0.20	1.24 ± 0.34
Knee, Maximum Negative Power[a]	−0.71 ± 0.40	−0.99 ± 0.43	−0.71 ± 0.40	−1.08 ± 0.49
Ankle, Maximum Positive Power[a]	3.64 ± 0.48	4.59 ± 0.16	3.43 ± 0.70	4.36 ± 0.93

Values are mean ± SD. Ant-post, anterior-posterior; GRF, ground reaction force; length in meter, velocity in meter per second, position in degrees, force, torque, and power in newton per kilogram, newton-meter per kilogram, and watts per kilogram, respectively. [a]Indicates load effect $p < 0.05$. [b]Indicates interaction between gender and load $p < 0.05$.

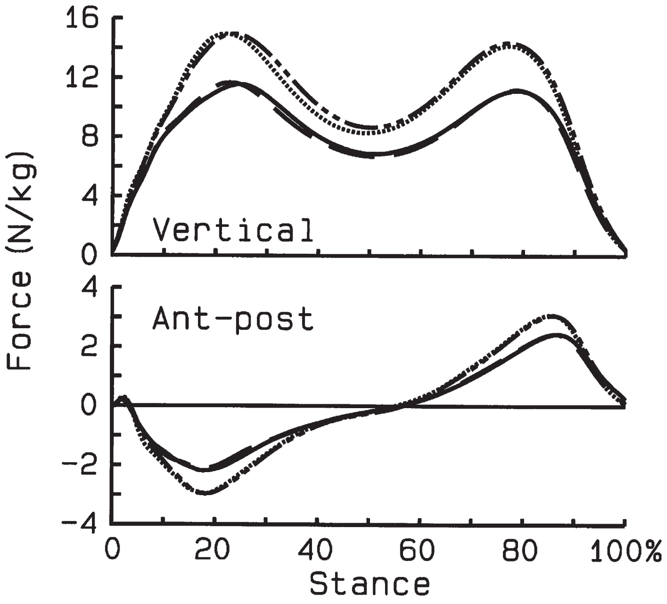

FIGURE 1. Vertical and anterior-posterior ground reaction forces (GRFs) during the stance phase for males (no load: dashed line; low load: dotted line) and females (no load: solid line; load: double-dashed line). The response to added load was statistically identical between genders, and males and females produced nearly identical mass-normalized GRFs in both the loaded and unloaded conditions. Maximum vertical GRFs, maximum braking, and maximum propulsive forces were increased in the loaded condition (all $p < 0.05$).

power in early stance ($p > 0.05$). Maximum mass-normalized hip positive power in late stance increased 15% in the loaded condition compared to unloaded condition ($p = 0.021$). Participants displayed increased mass-normalized knee extensor (65%, $p < 0.0001$) torque in the loaded condition compared to unloaded condition and maximum mass-normalized negative power in early stance (46%, $p < 0.0001$). The load contributed to a 23% increase in maximum mass-normalized ankle plantarflexor torque ($p < 0.0001$) and a 26% increase in positive power ($p < 0.0001$).

Correlational Analysis

Correlations performed between participant masses and forward trunk lean ($r = -0.63$), second maximum vertical GRF ($r = 0.94$), maximum ankle plantarflexor torque ($r = 0.87$), and maximum propulsive anterior GRF ($r = 0.82$) showed statistical significance using the following critical value for a two-tailed test at $p < 0.05$: (df = 9) = 0.60. These analyses showed that lighter participants carried the load with more forward lean but lower GRFs and ankle torque.

DISCUSSION

Interaction Between Gender and Load and Main Effect of Gender

We hypothesized an interaction effect between gender and load carriage such that females compared to males would respond differently to a standardized, additional heavy load

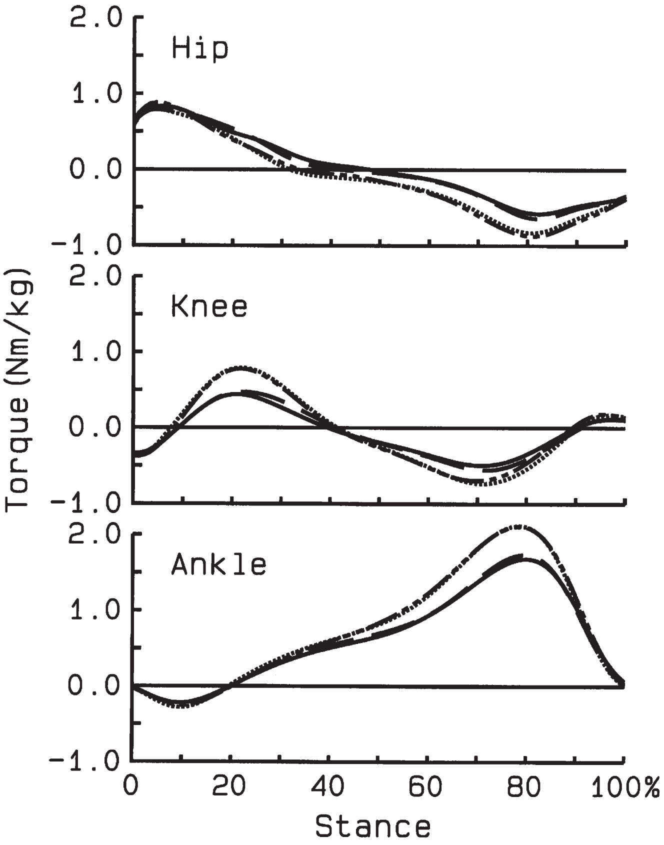

FIGURE 2. Sagittal plane joint torques during the stance phase for males (no load: dashed line; load: dotted line) and females (no load: solid line; load: double-dashed line). Positive values are extensor or plantarflexor, and negative values are flexor or dorsiflexor torques. The response to added load was statistically identical between genders, and males and females displayed statistically identical mass-normalized joint torques in the unloaded and loaded conditions. In the loaded conditions, participants had larger maximum mass-normalized hip flexor, knee extensor and flexor, and ankle plantarflexor torques compared to unloaded (all $p < 0.05$).

during level walking. An interaction was detected for average trunk position throughout the stance phase. Although both genders increased forward trunk lean with the load as seen previously,[8,34,35] females exhibited greater trunk lean in response to the load compared to males. The combination of the load and increased forward trunk lean likely created a flexor torque about the lower back, which required the counterbalancing action of the trunk extensor muscles[23] and also increased low-back loads,[7,36] particularly in the females. We suggest the seemingly small increase in trunk lean with load in females compared to males can be clinically meaningful over prolonged distances, which is known to accentuate load carriage effects.[34]

There were no gender differences detected for lower extremity gait kinetics or kinematics in the present study despite females being 5% shorter and having 8% less mass compared to males. The general similarity in gait between genders in the unloaded condition is in agreement with Kerrigan et al (1998) and Nigg et al (1994) who reported that

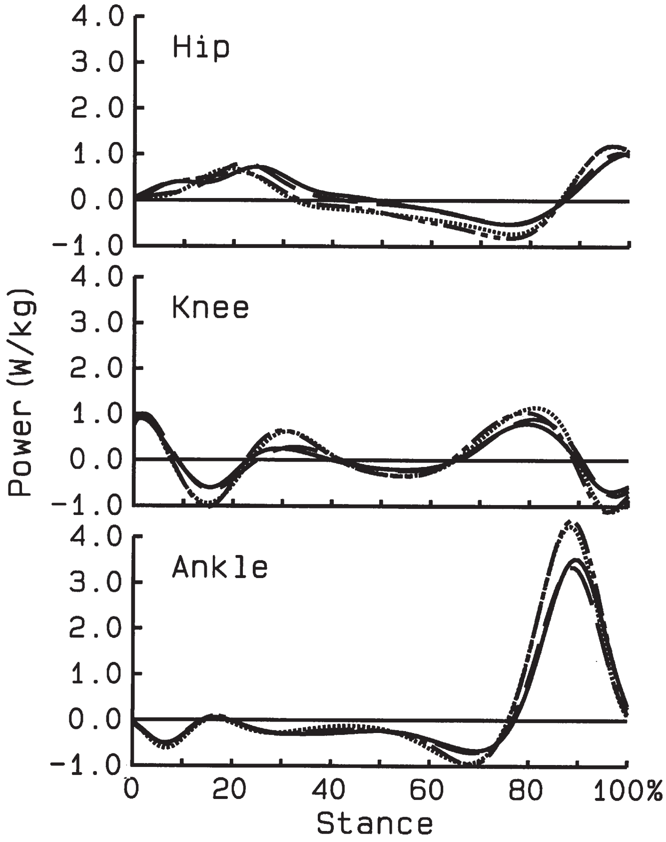

FIGURE 3. Sagittal plane joint power during the stance phase for males (no load: dashed line; load: dotted line) and females (no load: solid line; load: double-dashed line). Positive values indicate energy generation through concentric contractions, and negative values indicate energy absorption through eccentric contractions. The response to added load was statistically identical between genders, and males and females displayed statistically identical joint powers in the unloaded and loaded conditions. Load carriage increased maximum mass-normalized positive hip power in late stance, negative knee power in early stance, and ankle power in late stance (all $p < 0.05$).

there are more similarities than differences in gait biomechanics between males and females walking without a load.

When walking with a load normalized to body mass, males and females exhibited similar hip, knee, and ankle joint torques, and powers.[23] Thus, when loaded relative to body mass, females walked with identical mass-normalized gait biomechanics to those used by males. In the present study all participants carried the identical 22 kg standardized load with females being loaded more than males relative to their body mass, i.e., the load accounted for ~30% and ~32% of body mass for males and females, respectively. Surprisingly to us, the present standardized load for males and females yielded the same gender responses as the lighter loads used previously for females.[23] We anticipated that the differences in stature, body mass, and presumably muscle strength between males and females would yield gender differences in gait biomechanics when walking, as both small stature and female gender have been identified as risk factors for sustaining injuries, particularly stress fractures, in the military.[10,12,15,29]

We expected to observe females relative to males would shorten their stride more in response to the load partially to increase their stability[37] but mostly to attenuate the load-induced increase in external torques against their joints and muscles.[31] Finally, while reduced stride length does reduce external musculoskeletal torques, because of their smaller masses we still expected females to have larger mass-normalized joint torques and powers when carrying the heavy load compared to males. Since we did not observe these expected findings, the present results lead us to several new hypotheses on the higher injury rate of females compared to males in the military. It may be that the absence of gender differences in response to the heavy load may be a contributing factor to the increased injury rate observed in females. Namely, when carrying a load, females fail to adapt their gait mechanics in a manner that would account for their smaller stature and mass and presumably, lesser muscle strength compared to males. Indeed, gender may mask the actual injury causing factors: lower mass and strength that are directly related to load carriage ability.[38] In fact, more massive and stronger females were more successful at load carriage than smaller, weaker females.[39,40] As an initial exploration of this idea we correlated participant mass with several gait variables and we observed that the amount of forward trunk lean was inversely related to mass and the magnitude of the second maximum vertical GRF was directly correlated with mass. Thus, lighter people leaned more per unit body mass and pushed less against the floor compared to heavier people.

Main Effect of Load

The large increase in trunk flexion throughout the stance phase in the loaded compared to unloaded condition is consistent with the typical responses observed at the trunk when wearing a load concentrated primarily on the back.[8,19,35] The addition of the load on the back causes increased forward lean of the trunk and consequently increased flexion at the hip in an effort to maintain the center of mass position above the base of support. Surprisingly participants in the present study did not display increased hip extensor torque in the loaded conditions, which is in contrast to Silder et al[23] and Harman et al.[8] The load amount of 22 kg may not have been heavy enough to produce significant changes in hip extensor torque despite the large increase in hip flexion with added load. Harman et al[8] found that hip extensor torque was not different between a 6 kg and 20 kg or 33 kg load carried on the back, but was different from 6 kg to 47 kg. Similarly, Seay et al[41] found no differences in hip extensor torque between no load and a 15 kg vest, but did find differences in hip extensor torque between no load and a 55 kg vest. Seay et al[41] suggested that the lack of response at the hip when carrying a light load may indicate that the knee joint is responsible for the primary adjustment in response to loading. Admittedly we may have a type II statistical error in this case. The maximum hip torque from the loaded condition was larger than the unloaded condition; the observed F-ratio of 3.76 approached the statistically significant level of 5.32, and movements with greater hip flexion are known to have larger hip extensor torques.[42,43] Additionally, our participants displayed increased mass-normalized knee extensor and ankle plantarflexor torques in response to the load as did previous work,[23,41] which may help to explain the high rates of knee and ankle injuries in the military,[44,45] particularly in individuals who carry heavy loads.

Limitations

It is possible that the load amount of 22 kg was not heavy enough to cause significant gender and interaction effects in level walking. We also acknowledge that gender differences may be apparent after prolonged load carriage, which is known to accentuate load effects[34] but not in the immediate state as measured presently. Additionally, the differences in mass and height between male and female groups may have been too small to elicit significant gender and interaction findings. Lastly, the small sample size may have resulted in type II statistical error.

Conclusion

Although we found a significant interaction between gender and load for average trunk position, as well as numerous and substantial load effects in kinetic and energetic variables, we did not find gender effects or gender by load interaction effects in the kinetic and energetic variables in response to carrying a heavy, 22 kg load in a rucksack during level walking. The lack of gender differences may indicate that females do not adapt their gait mechanics to account for their smaller stature and lesser muscle strength compared to males, which may be a contributing factor to the high injury rate seen in female military recruits.

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