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The Effect of Military Load Carriage on Postural Sway, Forward Trunk Lean, and Pelvic Girdle Motion

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ABSTRACT

International Journal of Exercise Science 10(1): 25-36, 2017. Musculoskeletal injuries are a common occurrence in military service members. It is believed that the load carried by the service member impedes stability and alters back and pelvis kinematics, increasing their susceptibility to musculoskeletal injuries, specifically in the lower extremities. The purpose of this study was to examine the effects of two different loads on postural sway, forward trunk lean, and pelvic girdle motion in United States Army Cadets. Twenty male Army Reserve Officers' Training Corps Cadets participated in this study. Each participant performed the Modified Clinical Testing of Sensory Interaction (mCTSIB) Protocol and the Unilateral Stance (ULS) Protocol under three different rucksack load conditions (unloaded, 16.0 kg, and 20.5 kg loads). Mean postural sway velocity was recorded along with 2-D kinematics of the trunk in the sagittal plane and the pelvis in the frontal and sagittal planes. External loads of 16.0 kg ($p < 0.001$) and 20.5 kg ($p \leq 0.003$) significantly increased mean sway velocity by 16% to 52% depending on stance and visual condition, but did not produce significant changes in trunk and pelvic kinematics.

KEY WORDS: Spine, balance, stability, motion analysis, ARMY

INTRODUCTION

Service members are often required to carry heavy loads for prolonged periods during both training and combat (1, 21). Loads may include the Modular Lightweight Load-carrying Equipment System (MOLLE), Army Combat Helmet, rucksack and equipment, weapon, and body armor such as the improved outer tactical vest (IOTV). The maximum recommended weight of these loads varies depending on whether the load is carried for fighting (21.77 kg) or approach marching (32.66 kg) (1). Although the rucksacks are critical to mission success, the rucksack load can impede stability (e.g. balancing) (22) and movement (e.g. marching or running) (16),making it more difficult to balance and stop or initiate movement. This may necessitate greater torques at hip and trunk to control motion which has been shown to result in alterations to postural control (16, 22), requiring greater effort to control the load and maintain stability (29). The maintenance of stability may be further challenged in environments of low visibility while wearing a load carriage system. Service members are often required to carry rucksacks in conditions of low visibility (e.g. night missions or low visibility due to dust and sand). The collective impact of visual condition and load carriage (specifically rucksack usage) on service member's stability has received minimal attention. The ability to maintain stability while carrying a load within environments that obscure visual feedback may provide further insight into the prevalence of musculoskeletal injury to the spine (15, 28, 35).

Load induced alterations in postural control are often accompanied by compensatory kinematic changes at the trunk (2, 7, 11) and sagittal plane pelvic angle (2, 12, 30). Compensatory strategies may include adopting a more forward leaning trunk posture (4, 7, 11) and anteriorly tilted pelvis (6, 30) in order to help stabilize the body's center of mass (4). This adaptation may increase the torque on the lower back (21) and place greater stress on the low back muscles and vertebral disks (4).

Improved understanding of the biomechanical changes associated with increased load carriage may provide information needed to develop prevention protocols to help minimize injury, lost training/work days and medical costs due to injury in populations involved in regular load carriage. Past work available in this area has focused on experienced service members (4). However, musculoskeletal injuries to those with less load carriage experience are very common (18, 19, 26, 27), and biomechanical accommodations may differ in those with less experience carrying loads as compared to experienced service members. Musculoskeletal Injuries to those new to load carriage are concentrated in the lower extremities and back (19). One review (19) found that in male Army and Marine recruits, lower extremity and back injuries accounted for 24.7% to 76.7%. Many attribute these injuries to compensatory strategies elicited by the body in response to carrying loads (4, 6, 32).

Information identifying the biomechanical adaptations in response to load carriage may assist in the development of new standards and procedures for less experienced service members. Thus, the purpose of this study was to examine the effect of three different loads on postural sway, forward trunk lean, and pelvic girdle motion in Army Reserve Officers' Training Corps (ROTC) Cadets under two visual conditions. We hypothesized that increasing the load in less-experienced military personnel would significantly increase postural sway. Second, we hypothesized that increasing the load would produce significant increases in kinematic adaptations at the trunk and pelvic girdle. Finally, we hypothesized that increases in load would increase postural sway and kinematics significantly more under an eyes closed condition than an eyes open condition.

METHODS

Participants

Twenty male Army ROTC Cadets (age 21 ± 1.1 yrs.; mass 78.8 ± 9.1 kg; height 184.2 ± 6.1 cm) participated in this study. All participants were healthy, injury free and had prior experience carrying a rucksack. Exclusion criteria included any prior military service, surgery within the past year, acute injury, or any condition that may prevent successful completion of the tasks. All participants read and signed a consent form that was approved by the university institutional review board.

Protocol

The Basic Balance Master® (NeuroCom® International, INC., Clackamas, OR, USA) was used to evaluate postural control. The Basic Balance Master® (NeuroCom® International, INC., Clackamas, OR, USA) measures postural control using a portable force plate connected to a computer. For the NeuroCom® protocols (NeuroCom® International, INC., Clackamas, OR, USA) used in the current project (described in the procedures), the system uses center of gravity (COG) sway velocity (degrees·second^{-1}) as the measure of postural control. The system estimates the position and displacement of the COG, using a simple inverted pendulum approximation from the participant's height and sampled center of pressure data (14). The system calculates the COG sway velocity (i.e. postural sway velocity) by taking the ratio of the distance travelled by the COG (in degrees) to the time of trial (9, 14). Lower values indicate greater control (i.e. better balance) (9, 14). The Basic Balance Master® (NeuroCom® International, INC., Clackamas, OR, USA) is a common device used by healthcare practitioners to evaluate postural control in a clinical setting (34). The device has been shown reliable for use with various populations (10, 14) including healthy adults (31).

Kinematics were recorded using four digital camcorders. Two cameras [Canon 3CCD Digital Video Camcorder GL2 NTSC (Canon, INC., Japan)] were positioned in the frontal plane to record the motion of the retro-reflective markers. Additionally, two cameras [one additional Canon and one JVC mini DV (Victor Company of Japan, Limited, Malaysia)] were positioned in the sagittal plane. In each plane (i.e. frontal or sagittal), one camera recorded the entire body, while the other recorded the motion at the pelvic region. All cameras recorded at a frame rate of 60 fps with a video resolution of 720 p. Reflective markers were placed at the acromion process of the scapula, greater trochanter and lateral femoral condyle of the femur, lateral malleolus of the tibia, and the base of the 5th metatarsal. A hip belt apparatus was constructed and used to attach reflective markers that indicated the position of the right and left anterior superior iliac spines (ASIS). The belt was an adjustable aluminum apparatus that allowed for easier detection and measurement of ASIS movement (Figure 1).

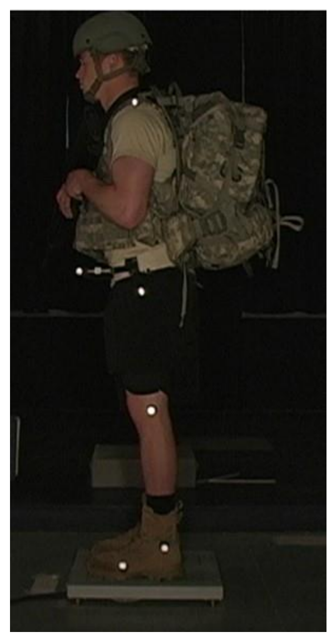

Figure 1. Sagittal view (Dartfish®)

Each participant reported to the Sport Biomechanics Laboratory for one testing day, lasting approximately 90 minutes. Participants reported to the testing session in shorts and shirt, and Army-issued boots. All participants completed a preliminary medical questionnaire, questionnaire on military and load carriage experience, and read and signed a consent form approved by the University's Institutional Review Board.

Height, weight and leg dominance were measured and recorded. Leg dominance was determined by gently pushing the participant from behind and recording the foot used to step forward (17). The dominant limb was used for all single leg measures. The participant was then fitted with the gear (per the Army Field Manual 21-18) (1) to be worn during testing of the 16.0 kg or 20.5 kg loads. The researchers provided and packed all gear to ensure consistency. Gear included the Modular Lightweight Load-carrying Equipment System (MOLLE), combat helmet, and a mock rifle. The participant held the mock rifle above the waist, barrel pointing to the ground at about a 45-degree angle. The rucksack weighed 16.0 kg or 20.5 kg, in accordance with the weights carried during the final ruck marches in Basic Combat Training and One Station Unit Training of entry-level Army Soldiers.

The kinematic and postural sway velocity data were collected simultaneously. Synchronization of the kinematic and postural sway velocity data was not possible due to the limitations of the technology. We collected postural sway velocity data on the Basic Balance Master® (NeuroCom® International, INC., Clackamas, OR, USA) using two NeuroCom® testing protocols: Modified Clinical Testing of Sensory Interaction and the Unilateral Stance tests protocols (NeuroCom® International, INC., Clackamas, OR, USA) using Balance Master®). Each participant performed the Modified Clinical Testing of Sensory Interaction Protocol. The protocol requires two visual conditions (eyes open and eyes closed). The participants performed the Modified Clinical Testing of Sensory Interaction Protocol with and without the eyes open for each of the load conditions (unloaded, 16.0 kg load, and 20.5 kg load) using a double leg stance on a firm surface. The participants completed both an eyes open and eyes closed visual condition for each load before progressing to the next load condition. For each load, the eyes open and eyes closed conditions were separated by a 45-second rest period. Each of the visual conditions lasted for a 60-second period. Between the 16.0 kg load and 20.5 kg load, all the gear was removed and 4.55 kg was added to the rucksack. The participants were given 2 minutes of rest between load conditions (Figure 2A). The participants performed six balance tests for this protocol.

For the Modified Clinical Testing of Sensory Interaction Protocol, the participant was placed onto the force platform in a forward facing position, with the participant's feet centered on the force plate. To center the participant's feet on the force plate, the medial malleolus of each foot was centered directly over the horizontal line on the force plate, while the lateral aspect of the calcaneus of each foot was position to the Short (S), Medium (M) or Tall (T) line indicated on the force platform. The position of the lateral calcaneus was based on participant height: S (76 cm to 140 cm), M (141 cm to 165 cm) and T (166 cm to 2013 cm). Participants were instructed to fix their gaze on a mark on the wall positioned at eye level during all eyes open balance tests to unify visual input and minimize the effect on balance (8, 13).

The Unilateral Stance Protocol required participants to complete two balance tests (one with the eyes open and one with the eyes closed) per load condition (unloaded, 16.0 kg load and 20.5 kg load) using a single leg stance on a firm surface for a total of six tests for the protocol. Each balance test was 60 seconds in duration, with a 45-second rest between balance test and a 2-minute rest between each load condition. The load progression and testing order were identical to the prior protocol. All gear was removed between the 16 kg and 20.5 kg load conditions (Figure 2B).

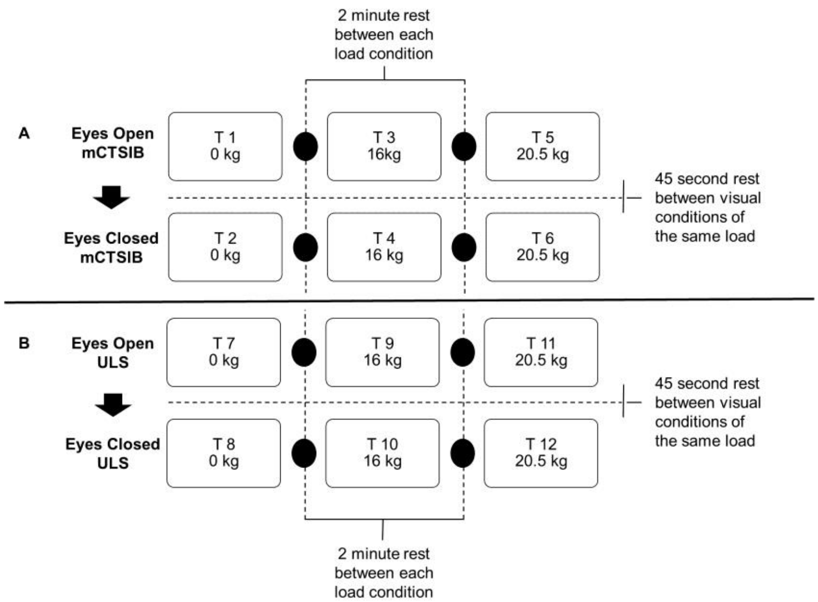

Figure 2. Diagram of testing procedures; A) Modified Clinical Testing of Sensory Interaction (mCTSIB) Protocol and B) Unilateral Stance (ULS) Protocol (note: T= test order)

The Unilateral Stance Protocol required participants to stand on the dominant limb with the non-weight bearing leg bent at the knee between 45 and 90 degrees of flexion. Participants gently touched down with the free foot if they felt they were falling. Each touchdown of the free foot was counted and recorded. A trial was considered a failure if the participant completely stepped off the platform or their foot moved from the appropriate spot on the platform. The literature indicates that for ages 18-36 postural sway velocity values for the eyes open and eyes closed conditions range from $0.7 \pm 0.2° \cdot s^{-1}$ to $1.16 \pm 1.25° \cdot s^{-1}$ (3, 23) and $3.96 \pm 2.52° \cdot s^{-1}$ to $6.42 \pm 3.81° \cdot s^{-1}$ (3), respectively, depending on the tested limb (right or left) and foot posture (normal, prone or supine) for the Unilateral Stance test.

Dartfish Live® (Dartfish, Switzerland) was used to determine trunk and pelvic angles under each load (0 kg, 16.0 kg and 20.5 kg) at the start and completion of each 60-second stance period. Previous research has found that Dartfish® is a valid reliable 2-D analysis software for measuring joint kinematics (20, 24). Forward trunk lean angles were measured in the sagittal plane and calculated by measuring the angle formed between two lines: the vertical reference line and a line running between the greater trochanter and acromion process (Figure 1). The marker representing the greater trochanter was positioned directly on the vertical reference line prior to calculating the angle. The change in angle was determined by finding the difference between the starting and ending angles for each test at the start and finish of the 60-second period.

The ASIS (indicated by the marker on the hip belt apparatus) of the dominant limb was positioned directly on a horizontal referenced line in the sagittal (anterior-posterior) or frontal plane (medial-lateral) to calculate the start and end angles for each test for both the anterior-posterior (Figure 1) and medial-lateral pelvic angles (Figure 3). The change in angle for the anterior-posterior and medial-lateral pelvic kinematics was determined by finding the difference between the starting and ending angles for each test.

Figure 3. Frontal view (Dartfish®)

Statistical Analysis
Data were analyzed using statistical software (SPSS 19, IBM Corp., Chicago, IL, USA). Four, 3 x 2 x 2 (load x stance x eyes) repeated measures ANOVAs with post-hoc comparisons using Boneferroni's adjustments were completed to analyze the postural control data and the three kinematic data sets (forward trunk lean, anterior-posterior pelvic tilt, and medial-lateral pelvic tilt). Alpha level was set a priori at $\alpha \leq 0.05$.

RESULTS

We identified no significant three-way interaction (load x visual conditions x stance) effects for sway velocity, forward trunk lean, or in the anterior-posterior pelvic motion and medial-lateral pelvic motion. A significant two-way interaction was observed for stance x eyes and load x eyes for mean sway velocity. Post-hoc analysis revealed significant main effects on postural sway velocity. In a double leg stance, mean postural sway velocity significantly increased between the 0 kg and 16 kg ($p < 0.001$), 0 kg and 20.5 kg ($p < 0.001$), and 16 kg and 20.5 ($p = 0.026$) loads. In a single leg stance, mean postural sway velocity significantly increased between the 0 kg and 16 kg ($p < 0.001$) and 0 kg and 20.5 kg ($p = 0.003$) loads. There was not a significant difference ($p = 0.648$) in mean postural sway velocity between the 16 kg and 20.5 kg loads while in a single leg stance. Table 1 summarizes the effect stance and vision had on mean postural sway velocity under the three load conditions.

Table 2 summarizes the effects of load on forward trunk lean, anterior-posterior pelvic girdle rotation and medial-lateral pelvic girdle rotation. Forward trunk lean, anterior-posterior pelvic girdle rotation, or medial-lateral pelvic girdle rotation were not significantly altered by load. Significant two-way interactions were observed for stance x eyes for forward trunk lean, anterior-posterior pelvic girdle rotation or medial-lateral pelvic girdle rotation. Post-hoc analyses identified significant main effects of vision (eyes open vs. eyes closed) and stance (double leg vs single leg) on forward trunk lean, anterior-posterior pelvic girdle rotation and medial-lateral pelvic girdle rotation.

Table 1. Mean postural sway velocity [degrees second^{-1} (standard deviation)] for balance under each condition

Stance	Visual condition	Rucksack weight	Postural sway velocity ($°\cdot s^{-1}$)	p-value^{a}	p-value^{b}	p-value^{c}	p-value^{d}
DL	EO	0 kg	0.27 (0.07)	--	0.002	<0.001	<0.001
		16 kg	0.34 (0.08)	--	<0.001	<0.001	<0.001
		20.5 kg	0.41 (0.15)	--	<0.001	<0.001	<0.001
DL	EC	0 kg	0.38 (0.17)	<0.001	--	<0.001	<0.001
		16 kg	0.49 (0.13)	<0.001	--	<0.001	<0.001
		20.5 kg	0.52 (0.17)	<0.001	--	<0.001	<0.001
SL	EO	0 kg	1.03 (0.15)	<0.001	<0.001	--	<0.001
		16 kg	1.19 (0.17)	<0.001	<0.001	--	<0.001
		20.5 kg	1.27 (0.40)	<0.001	<0.001	--	<0.001
SL	EC	0 kg	2.20 (0.71)	<0.001	<0.001	<0.001	--
		16 kg	2.69 (0.82)	<0.001	<0.001	<0.001	--
		20.5 kg	2.83 (0.93)	<0.001	<0.001	<0.001	--

DL=double leg; SL=single leg; EO=eyes open; EC=eyes closed; The letters 'a', 'b', 'c', and 'd' represent significant differences; a=DL EO; b=DL EC; c=SL EO; d=SL EC

Table 2. Mean initial angular position in degrees (standard deviation) and change in position in degrees (standard deviation) for kinematic motions under each condition (0, 16 and 20.5 kg)

Motion	Visual condition	Stance	Rucksack weight	Initial angle(°)	Change	Motion	Visual condition	Stance	Rucksack weight	Initial angle(°)	Change
AP	EO	DL	0 kg	-8 (7)	0 (2)	ML	EO	SL	0 kg^{d,e*}	2 (4)	-2 (3)
			16 kg	-10 (8)	0 (2)				16 kg	2 (3)	-2 (3)
			20.5 kg^{b*}	-13 (8)	1 (2)				20.5 kg^{f*}	3 (4)	-2 (2)
AP	EC	DL	0 kg	-8(7)	0 (2)	ML	EC	SL	0 kg^{d*}	2 (3)	0 (3)
			16 kg^{a*}	-11 (9)	0 (2)				16 kg	2 (3)	0 (3)
			20.5 kg^{c*}	-13 (8)	1 (8)				20.5 kg^{f,g*}	2 (3)	1 (4)
AP	EO	SL	0 kg	-10 (8)	-1 (4)	FTL	EO^{i*}	DL^{h*}	0 kg	-2 (4)	0 (2)
			16 kg	-12 (8)	-1 (5)				16 kg	3 (2)	0 (1)
			20.5 kg^{b*}	-15 (10)	4 (9)				20.5 kg	4 (3)	0 (2)
AP	EC	SL	0 kg	-12 (8)	-3 (4)	FTL	EC^{i*}	DL	0 kg	-2 (4)	0 (1)
			16 kg^{a*}	-14 (10)	-3 (5)				16 kg	2 (2)	0 (1)
			20.5 kg^{c*}	-16 (11)	-3 (7)				20.5 kg	4 (2)	1 (1)
ML	EO	DL	0 kg^{e*}	-2 (2)	0 (1)	FTL	EO	SL^{h*}	0 kg	0 (4)	0 (1)
			16 kg	-1 (3)	0 (1)				16 kg	3 (2)	0 (2)
			20.5 kg^{g*}	-1 (2)	0 (1)				20.5 kg	4 (3)	2 (4)
ML	EC	DL	0 kg	-2 (2)	0 (1)	FTL	EC	SL	0 kg	0 (3)	2 (4)
			16 kg	-1 (2)	0 (2)				16 kg	4 (3)	3 (3)
			20.5 kg	-1 (3)	0 (1)				20.5 kg	6 (4)	2 (4)

Note. AP, anterior-posterior pelvic motion; ML, medial-lateral pelvic motion; FTL, forward trunk lean; DL, double leg; SL, single leg; EO, eyes open; EC, eyes closed; Paired letters (a,b,c,d,e,f,g,h,i) indicate significant main effects between the paired letters.*$p \leq 0.05$

DISCUSSION

The purpose of this study was to examine the effect of three different loads on postural sway, forward trunk lean, and pelvic girdle motion in Army Reserve Officers' Training Corps (ROTC) Cadets under two visual conditions. The primary finding was that external posterior loads of 16.0 kg and 20.5 kg produced changes in mean postural sway velocity over a 60-second stance period in Army ROTC Cadets. The same external loads did not result in significant kinematic adaptations over the 60-second stance period.

We hypothesized that increasing the load would significantly increase postural sway. In the present study, mean postural sway velocity during a double leg stance increased from $0.27°\cdot s^{-1}$ to $0.34°\cdot s^{-1}$ (16.0 kg load) and $0.41°\cdot s^{-1}$ (20.5 kg) under the eyes open condition and from $0.38°\cdot s^{-1}$ to $0.49°\cdot s^{-1}$ (16.0 kg load) and $0.52°\cdot s^{-1}$ (20.5 kg) with the eyes closed. Mean postural sway velocity during single leg stand with eyes open increased from $1.03°\cdot s^{-1}$ to $1.19°\cdot s^{-1}$ (16.0 kg load) and $1.27°\cdot s^{-1}$ (20.5 kg); and from $2.20°\cdot s^{-1}$ to $2.69°\cdot s^{-1}$ (16.0 kg load) and $2.83°\cdot s^{-1}$ (20.5 kg) during the eyes closed condition. The Army Field Manual 21-18 (1) indicates that after proper training experienced soldiers should be expected to carry loads up to 45% of their body weight on a ruck march. New soldiers begin load carriage with lighter weights. The 16 kg and 20.5 kg loads used in the present study would have represented 20% and 26% of our participants' average mean body mass which is more representative of training weights. Future research using a load at the 45% of body weight may produce greater changes.

Previous studies on load carriage have used other measures of postural control, making a direct comparison of our results difficult. However, we did find that investigators using other metrics to quantify postural control (center of pressure) reported that load significantly influence balance (16, 29). One group (16) observed that anterior–posterior center of pressure excursion and medial–lateral excursion increased as much as 54% and 131%, respectively, in college aged females when load conditions went from unloaded to a 18.1 kg load. In the second study (29) investigators observed that center of pressure excursion using load conditions of 6 kg, 16 kg and 40 kg with enlisted soldiers. The investigators (29) observed that, compared with the 6-kg load, the 16- and 40-kg loads increased anterior–posterior center of pressure excursion by 21% and 42%, respectively. The group (29) also reported increases of 13% and 56% for the 16- and 40-kg loads, respectively, when compared to 6 kg for medial–lateral center of pressure excursion. In both these studies, the researchers had the participants stand in a double leg stance with their eyes open. In the present study, we observed that in a double leg stance with the eyes open mean postural sway velocity also increased significantly. Together with previous research, it appears that external loads greater than or equal to 16 kg result in substantial alterations to postural control.

The present study also examined kinematic changes (i.e. adaptations across the 60-second stance period) at the trunk and pelvic girdle due to external load carriage. We hypothesized that, as the external load increased there would be significant kinematic adaptations at the trunk and pelvic girdle. However, our results revealed that increasing the external loads did not result in significant kinematic changes over the 60-second stance duration any of the 12 balance tests for forward trunk lean, anterior-posterior pelvic motion, or medial-lateral pelvic motion. This is perhaps due to the sensitivity of our analysis software. In the current study, kinematics were analyzed using the Dartfish Live® analysis software. Dartfish Live® analysis software is a useful tool for analyzing changes in kinematics. However, it may lack the sensitivity to measure smaller angular changes, such as those observed at the trunk and pelvic girdle in the present study. The use of a more sensitive instrumentation such as 3-D motion capture would help provide a clearer insight into the influence of load on trunk and pelvic kinematics during a quiet stance. Investigators (4) using 3-D motion capture reported observing male soldiers (with an average body mass of 74.9 kg) soldiers leaned forward at the trunk significantly while walking with loads of 39.95 kg and 50.05 kg compared to loaded conditions under 15.96 kg. Investigators (30) have also observed that anterior pelvic motion was significantly altered with loads as low as 15% of body mass in college-age females while walking. Future studies should be performed using 3-D motion capture to validate our results in regards to the influence of incremental loading on trunk and pelvic motion during single and double leg standing.

Ruck marches are often completed in low visibility conditions; thus, in the present study cadets performed tasks with the eyes open and closed. Single legged stance was included to better replicate a complete gait cycle (compared to double legged standing). In the single leg stance, significant alteration in forward trunk lean and anterior-posterior pelvic motion with the eyes closed were observed. Further research is required to understand the interplay of load and visual conditions. A better understanding of the interplay of load and visual conditions will help in the development of effective training interventions.

This study has several limitations. Reflective markers were applied to the skin and clothing above bony processes near a joint. Movement of the skin, clothing or of the hip belt apparatus could have produced errors (5, 25). However, with the exception of the marker placement on the greater trochanter, this method has been validated in other studies (33). Dartfish® software has limitations when measuring joint angle changes that are ≤6° which could have impacted our kinematic findings. We did not assess balance touchdowns in the data analysis, potentially leading to deflated results of the influence of load on postural sway. In addition, participants in this study had some experience with load carriage as opposed to Initial-Entry Trainees who would have no load carriage experience.

This study suggests that increased load using military load carriage systems influences postural sway in male Army ROTC Cadets with little load carriage experience. A 16.0 kg to 20.5 kg load increased all postural sway measures in double leg and single leg stances. Limiting the participants' ability to utilize visual feedback strongly influenced the participants' postural sway velocity.

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