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# Effects of load carriage on biomechanical variables associated with tibial stress fractures in running

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ABSTRACT

*Background:* Military personnel are required to run while carrying heavy body-borne loads, which is suggested to increase their risk of tibial stress fracture. Research has retrospectively identified biomechanical variables associated with a history of tibial stress fracture in runners, however, the effect that load carriage has on these variables remains unknown.
*Research question:* What are the effects of load carriage on running biomechanical variables associated with a history of tibial stress fracture?
*Methods:* Twenty-one women ran at 3.0 m/s on an instrumented treadmill in four load carriage conditions: 0, 4.5, 11.3, and 22.7 kg. Motion capture and ground reaction force data were collected. Dependent variables included average loading rate, peak absolute free moment, peak hip adduction, peak rearfoot eversion, and stride frequency. Linear mixed models were used to asses the effect of load carriage and body mass on dependent variables.
*Results:* A load x body mass interaction was observed for stride frequency only ($p = 0.017$). Stride frequency increased with load carriage of 22.7-kg, but lighter participants illustrated a greater change than heavier participants. Average loading rate ($p < 0.001$) and peak free moment ($p = 0.015$) were greater in the 22.7-kg condition, while peak rearfoot eversion ($p \leq 0.023$) was greater in the 11.3- and 22.7-kg conditions, compared to the unloaded condition. Load carriage did not affect peak hip adduction ($p = 0.67$).
*Significance:* Participants adapted to heavy load carriage by increasing stride frequency. This was especially evident in lighter participants who increased stride frequency to a greater extent than heavier participants. Despite this adaptation, running with load carriage of ≥11.3-kg increased variables associated with a history of tibial stress fracture, which may be indicative of elevated stress fracture risk. However, the lack of concomitant change amongst variables as a function of load carriage may highlight the difficulty in assessing injury risk from a single measure of running biomechanics.

## 1. Introduction

In the military, incidence rates of stress fracture have been reported as high as 31 % [1], with the tibia being one of the most common sites of fracture, accounting for 14–74 % of stress fractures [2–4]. In both recreational and military populations, women are more than twice as likely to experience a stress fracture than men [4,5], and this discrepancy may be explained, at least in part, by differences in gait mechanics [6]. Indeed, women run with increased peak hip adduction, peak hip internal rotation, peak knee abduction, and peak rearfoot eversion when compared to men [7–10]. Interestingly, both peak hip adduction and peak rearfoot eversion have been identified as risk factors for tibial stress fractures [11,12].

It has also been reported that women with a history of tibial stress

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fracture run with greater average vertical loading rates and peak absolute free moment compared to non-injured controls [11,12]. The association of these variables with a history of stress fracture suggests they are in some way related to bone loading. In support of this statement, Yang et al. [13] reported fair to moderate correlations ($r^2$ = 0.25-0.66) between the free moment and tibial torsion (i.e., the angle of twist of the proximal tibia relative to the distal tibia). While this demonstrates that the free moment may act as a surrogate measure of tibial torsion, the remaining variables have not been correlated to *in vivo* tibial mechanics (i.e., stress, strain, deformation). Of course, this does not discredit the examination of these variables, but it may limit the ability to make statements about the causal relationship between these variables and tibial stress fracture.

Running with body-borne loads (a.k.a. load carriage running) is necessary in military operations, and long-distance load carriage running is frequently employed to maintain fitness levels in Soldiers [14]. While it is known that load carriage increases the risk of stress fracture development [15], little is known about how load carriage running alters biomechanical variables that are associated with a history of tibial stress fracture. Loading rate has been shown to increase with load carriage, but Lobb et al. [16] only examined heavy load carriage magnitudes (≥20 kg), with no reference to unloaded running, which makes it difficult to interpret the magnitude of the observed changes. Previous studies have shown that hip adduction and rearfoot eversion does not change during load carriage running, but these studies included both men and women [17], or have used relatively lighter load carriage masses [18]. Including both men and women may mask the effects of load carriage due to differences in running biomechanics [8], and light load carriage mass would presumably reduce the biomechanical demands of the running task. Furthermore, many studies of load carriage biomechanics use body-borne loads that are a percentage of the participant's body mass [17–19]. This does not accurately reflect the nature of load carriage in the military, where individuals are required to carry an absolute load that is irrespective of their body mass [20]. Intuitively one would expect that body-borne loads that are a greater percentage of an individual's body mass would result in greater biomechanical changes. Thus, more work is needed to specifically examine the effect of load carriage running on biomechanical variables associated with tibial stress fractures in women with differing body stature.

The purpose of this study was to examine the effect of load carriage and body mass on biomechanical variables previously associated with tibial stress fracture risk during running. To this end, we quantified average loading rate, peak absolute free moment, peak hip adduction and peak rearfoot eversion in physically active women running with body-borne loads of 0, 4.5, 11.3, 22.7 kg. We hypothesized that 1) biomechanical variables would increase concomitantly with body-borne loads, and 2) lighter women would exhibit greater biomechanical changes than heavier women for a given change in load carriage magnitude.

## 2. Methods

### 2.1. Participants

Twenty-one physically active women (19 ± 1 years, 164.04 ± 8.49 cm, 59.89 ± 6.32 kg) were enrolled in this study after providing written informed consent. All participants were required to be free from injuries limiting physical activity for the three months prior to data collection, were experienced treadmill runners, and were participating in physical activity at least three times per week. All aspects of the study were first approved by the University of Calgary Conjoint Health Research Ethics Board and by the Human Research Protection Office at the United States Army Medical Research and Development Command, Fort Detrick, MD.

### 2.2. Data collection

Twenty-one retroreflective markers were placed on each participant for motion capture analysis. This included thirteen anatomical landmark markers and eight segmental tracking markers. The anatomical landmark markers were placed on the left and right anterior superior iliac crest, posterior superior iliac crest, greater trochanters, right medial and lateral femoral condyles, medial and lateral malleoli, heel, $2^{nd}$ metatarsal and $5^{th}$ metatarsal. The segment tracking markers included clusters of four markers placed on the right anterior thigh, and on the right posterior shank.

A static motion capture trial was recorded prior to dynamic trials to determine joint centers; the greater trochanter, medial femoral condyle, and malleoli markers were subsequently removed. The ankle joint center was placed at 50 % of the distance between medial and lateral malleoli markers, knee joint center was placed at 50 % of the distance between medial and lateral femoral epicondyles, and hip joint center was placed at 70 % of the distance between the left and right greater trochanters. Anatomical coordinate systems were defined for the thigh and shank using femoral epicondyles and the hip joint center and the medial and lateral malleoli and femoral epicondyles, respectively. Marker coordinates were used to track motion of the pelvis, femur, shank, and foot through space, which were each modeled as rigid segments. During dynamic trials, force (2000 Hz) and motion capture data (200 Hz) (Vicon Motion Systems Ltd., Oxford, UK) were recorded synchronously while the participants ran on an instrumented treadmill (Bertec Corporation, Columbus, OH) at a speed of 3.0 m/s. Participants ran under four load conditions including: 0, 4.5, 11.3, 22.7 kg in a randomized order. The load magnitudes were chosen to reflect the training loads worn during basic combat training exercises in the United States Army [21]. Load conditions were accomplished using a weighted vest consisting of equally distributed combinations of 1.2 kg cast iron weights (V-Max, weightvest.com). This configuration was previously used to simulate body-borne load conditions similar to a military setting [22,23]. Each participant completed a 5-minute progressive warm-up to accommodate to the treadmill. For each load condition, participants were given 30 s to acclimate after achieving the prescribed speed. Data were collected for 20 s following acclimation, and between each trial, participants were permitted to rest as much as they desired.

### 2.3. Data analysis

Motion capture and ground reaction force data were analyzed using MATLAB software (v. R2016b, The MathWorks, Natick, MA). A 20-N threshold for the vertical ground reaction force was used to identify the stance phase of running; nine strides for the right leg were extracted for the analysis of each condition. Average loading rate was calculated as the average slope of the vertical ground reaction force between 20–80 % of the instant from heel strike to peak impact force [24]. The free moment (FM) was calculated from the force platform according to:

$$FM = M_z - ((COP_{AP} \times F_{ML}) + (COP_{ML} \times F_{AP}))$$

where $M_z$ is the moment about the vertical axis, $COP_{AP}$ and $COP_{ML}$ are the center of pressures in the anterior-posterior and medial-lateral directions, respectively, and $F_{AP}$ and $F_{ML}$ are the force in the anterior-posterior and medial-lateral directions, respectively. Joint kinematics were calculated using a flexion-extension, adduction-abduction, internal-external Cardan sequence of rotations. Ground reaction force data, hip adduction, and rearfoot eversion were filtered using a $4^{th}$ order Butterworth filter with a cut-off frequency that retained 95 % of the signal power [25]. The range of cut-off frequencies were 32–42, 3–9, and 6 − 29 Hz for the ground reaction force, hip adduction, and rearfoot eversion, respectively. Stride frequency was calculated using the time between successive ipsilateral foot contacts. Dependent variables included the average loading rate, peak absolute free moment,

**Table 1**
Beta estimates and lower and upper limits for 95 % confidence intervals for mixed model parameters. Only models where the interaction term was significant include interaction factors.

| | Estimate (β) | 95% CI Limits |
|---|---|---|
| **Average Vertical Loading Rate (BW/s)** | | |
| Intercept | 75.06 | 36.5, 113.7 |
| 4.5 kg | 2.17 | −0.4, 4.8 |
| 11.3 kg | 2.79 | 0.2, 5.4 |
| 22.7 kg | 10.38 | 7.8, 13.0 |
| Body mass | −0.40 | −1.0, 0.2 |
| **Peak Absolute Free Moment (Nm/BW*Height)** | | |
| Intercept | 22.64 | 14.1, 31.2 |
| 4.5 kg | 0.88 | $-0.1\text{x}10^{-1}$, 1.76 |
| 11.3 kg | 0.76 | −0.1, 1.64 |
| 22.7 kg | 1.53 | 0.7, 2.4 |
| Body mass | −0.23 | −0.4, −0.1 |
| **Peak Hip Adduction (°)** | | |
| Intercept | 12.00 | −5.9, 29.9 |
| 4.5 kg | 0.50 | −0.3, 1.3 |
| 11.3 kg | 0.18 | −0.6, 1.0 |
| 22.7 kg | 0.23 | −0.6, 1.0 |
| Body mass | −0.03 | −0.3, 0.3 |
| **Peak Rearfoot Eversion (°)** | | |
| Intercept | −12.33 | −25.8, 1.2 |
| 4.5 kg | −0.46 | −1.1, 0.1 |
| 11.3 kg | −0.86 | −1.5, −0.2 |
| 22.7 kg | −1.54 | −1.5, −0.2 |
| Body mass | $-0.04 \times 10^{-1}$ | −0.2, 0.2 |
| **Stride Frequency (strides/min)** | | |
| Intercept | 100.93 | 81.0, 120.8 |
| 4.5 kg | 2.92 | −3.7, 9.6 |
| 11.3 kg | 5.23 | −1.4, 11.9 |
| 22.7 kg | 13.97 | 7.3, 11.9 |
| Body mass | −0.23 | −0.6, 0.1 |
| 4.5 kg x body mass | −0.05 | −0.2, 0.1 |
| 11.3 kg x body mass | −0.06 | −0.2, 0.1 |
| 22.7 kg x body mass | −0.18 | −0.3, 0.1 |

peak hip adduction, peak rearfoot eversion, and stride frequency.

For each subject, dependent variables from the nine analyzed strides were averaged for each condition. Linear mixed models were used to evaluate the relationship between the dependent variables and the predictor variables (load and body mass). Load was treated as a categorical factor and body mass was treated as a continuous factor, while random intercepts were included for each participant. Statistical analyses were performed in R (R Core Team, 2018) within RStudio (Version 1.1.463, RStudio, Inc.), using *lme4, lmerTest,* and *multcomp* packages [26–29]. The load carriage by body mass interaction term was removed from models when a likelihood ratio test indicated a non-significant difference between models at $p < 0.05$. In the case of a significant main effect, pairwise comparisons were performed using a Bonferroni correction for multiple comparisons with an adjusted $\alpha = (0.05/4) = 0.0125$. All data, unless specified, are represented as mean ± standard deviation.

## 3. Results

Beta estimates and 95 % confidence limits for mixed model factors can be found in Table 1, while mean and standard deviation for variables of interest can be found in Table 2.

A load carriage by body mass interaction was observed for stride frequency (F(3,63) = 3.66, p = 0.017) only (Table 1&2). While stride frequency increased for all participants during the 22.7 kg condition, the effect was influenced by body mass, where each 1 kg increase in body mass was associated with a 0.18 strides/min reduction in stride frequency (β=-0.18, p = 0.004). Thus, lighter participants increased stride frequency more than heavier participants in the 22.7 kg condition. Load carriage had no effect on peak hip adduction (F(3,60) = 0.52, p = 0.67) (Fig. 1), but load carriage did affect average loading rates (F(3,60) = 22.69, p < 0.001), peak rearfoot eversion (F(3,60) = 8.39, p < 0.001), and peak free moment (F(3,60) = 3.80, p = 0.015) (Table 1). Average loading rates were greater in the 11.3 (β = 2.79, p = 0.04) and 22.7 kg (β = 10.38, p < 0,001) conditions, compared to 0 kg (Fig. 2). Peak rearfoot eversion was greater in the 11.3 (β=-0.86, p = 0.009) and 22.7 kg (β= -1.54, p < 0.001), compared to the 0 kg condition, (Fig. 3). Peak free moment was greater in the 22.7 kg condition (β = 1.53, p = 0.001) compared to 0 kg only (Table 1).

Body mass was observed to be a significant predictor for peak absolute free moment only (F(1,19) = 9.90, p = 0.05), where every 1 kg increase in body mass was associated with a 0.23 Nm/BW*Height reduction in peak absolute free moment (Table 1).

## 4. Discussion

The purpose of this study was to examine the effect of load carriage and body mass on biomechanical variables previously associated with tibial stress fracture risk during running

Stride frequency was the only variable to exhibit a load carriage by body mass interaction, where lighter participants increased their stride frequency more during 22.7 kg running than heavier participants. Increased stride frequency appears to be an adaptation made by participants to reduce the demands of load carriage running. Despite this adaptation, heavy load carriage (11.3 and 22.7 kg) was associated with increases in average loading rate, peak free moment, and rearfoot eversion. No difference in loading rate was observed in lighter load carriage conditions, and peak hip adduction was similar across all conditions. Although these findings do suggest an increased risk of stress fracture with heavy load carriage, the disparate effects of load carriage on the examined variables may highlight the difficulty in assessing stress fracture risk from single measures of running biomechanics.

Previous research investigating unloaded running reported that average loading rates was approximately 12 BW/s greater when comparing individuals with a history of tibial stress fractures to non-injured controls [11]. The present study observed increased loading rates in the 11.3 and 22.7 kg conditions, during which average loading rates were 2 and 10 BW/s greater than the 0 kg condition. Based on the small change in 11.3 kg it may be tempting to assume that there would be little to no risk of tibial stress fracture in this condition with repeated running. However, it has been observed that load carriage running with 11.3 kg increased tibial stress integral and the volume of highly stressed bone [18]. Thus, the results of Xu et al. [18] suggest that load carriage running of >11.3 kg may increase the risk of stress fracture, despite the minor changes observed in loading rate.

It is possible that only minor changes were observed in loading rate due to the increased stride frequency adopted by participants. Previous studies have demonstrated that decreasing stride length during running can reduce loading rates with and without body-borne loads, and in the present study, participants increased their stride frequency during the 22.7 kg condition by 3.9 %. However, the small change adopted by participants was insufficient to completely negate the effect of the 22.7 kg condition, as a 10 BW/s increase in loading rate was observed in this condition. If elevated loading rate is indicative of an increased stress fracture risk, it is possible that stress fracture risk is compounded by the increased stride frequency adopted by participants. Increasing stride frequency results in a greater number of loading cycles experienced by an individual, and in the presence of elevated tibial stress (as demonstrated by Xu et al. [18]), this would increase the risk of tibial stress fracture. This may be particularly important for smaller women, as they increased stride frequency more than larger women in the 22.7 kg condition.

Peak rearfoot eversion was also greater in both the 11.3 and 22.7 kg condition. However, peak rearfoot eversion for the 11.3 and 22.7 kg conditions were only 0.9° and 1.5° higher than the 0 kg condition, respectively. Although the observed changes in peak rearfoot eversion in the 25 lb and 50 lb condition were small (i.e., 0.9° and 1.5°,

**Table 2**
Mean (standard deviation) for discrete variables of interest. Superscript letters a,b,c,d denote significant difference from 0, 4.5, 11.3 and 22.7 kg condition, respectively, from *post hoc* testing of main effect of load carriage ($p \leq 0.005$).

| | 0 kg | 4.5 kg | 11.3 kg | 22.7 kg |
|---|---|---|---|---|
| **Average Vertical Loading Rate (BW/s)** | 50.9 (9.8) $^{d}$ | 53.1 (10.0) $^{d}$ | 53.7 (10.1) $^{d}$ | 61.3 (10.5) $^{a,b,c}$ |
| **Peak Absolute Free Moment (Nm/BW*Height)** * | 9.0 (2.9) $^{d}$ | 9.9 (3.3) | 9.8 (2.9) | 10.5 (1.9) $^{a}$ |
| **Peak Hip Adduction (°)** | 10.5 (4.4) | 11.0 (4.5) | 10.6 (4.6) | 10.7 (3.9) |
| **Peak Rearfoot Eversion (°)** | −12.6 (2.9) $^{c,d}$ | −13.1 (3.4) $^{d}$ | −13.5 (3.3) $^{a}$ | −14.1 (3.5) $^{a,b}$ |
| **Stride Frequency (strides/min)** | 86.9 (4.9) $^{d}$ | 87.1 (5.2) | 88.4 (5.2) | 90.3 (5.1) $^{a}$ |

* All peak absolute free moment values are x $10^{-3}$.

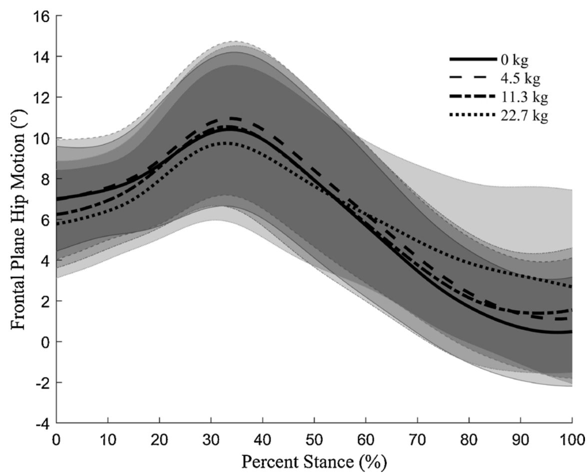


**Fig. 1.** Mean and standard deviation ensemble curves for frontal plane hip motion in each load carriage condition. Positive values indicate hip adduction.

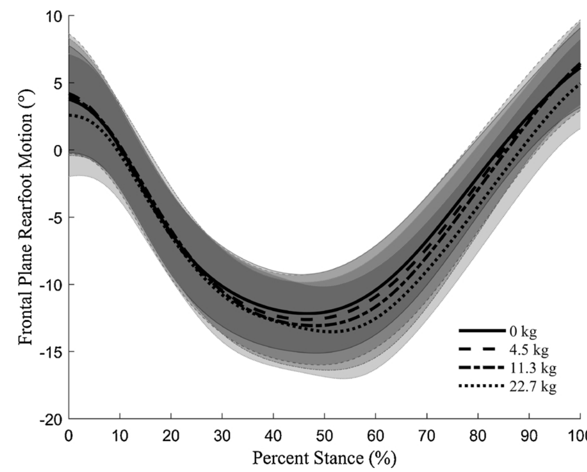


**Fig. 3.** Mean and standard deviation ensemble curves for frontal plane rearfoot motion in each load carriage condition. Negative values indicate eversion.

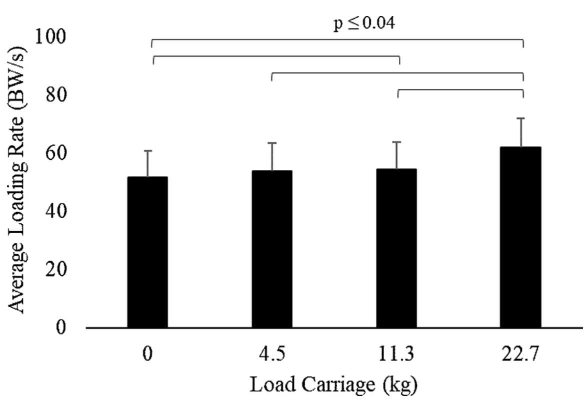


**Fig. 2.** Mean and standard deviation of average loading rate during all load carriage conditions.

respectively), Pohl et al. [11] observed a 2.7° increase in peak rearfoot eversion when comparing individuals with and without a history of tibial stress fracture. This suggests that small differences in foot/ankle movement patterns may influence the loading experienced by the tibia. In support of this statement, recent work suggests that small alterations in foot/ankle motion (subjects cued to exaggerate pronation) can have large effects on the torsional loading experienced by the tibia [30].

The free moment is a measure of the torque about the vertical axis between the foot and the ground during the stance phase of gait, and this measure is sensitive to foot pronation in running [31]. Given the increase in rearfoot eversion, a component of foot pronation, observed in the 11.3 and 22.7 kg conditions, one might expect the free moment to increase in these conditions as well. However, peak free moment was only greater during the 22.7 kg condition, so it is possible that the small (0.8°) change in rearfoot eversion in the 11.3 kg condition was not great enough to elicit a change in the free moment. In contrast, peak free moment was 1.5 Nm/BW*Height greater in the 22.7 kg condition than in the 0 kg condition. While less than the 3.0 Nm/BW*Height difference observed between women with and without a history of tibial stress fracture [32], these results may still suggest an elevated risk of tibial stress fracture during load carriage of 22.7 kg. Body mass was also observed to influence peak free moment, with heavier subjects running with a reduced free moment, across all conditions. This may be a function of normalising the free moment to body mass, as a mixed model constructed using a non-normalized free moment measure, detected no effect of body mass on peak free moment ($\beta = 0.087$, $p = 0.89$).

The present study observed no change in peak hip adduction between load carriage conditions, which verifies the findings of previous studies [17,18]. While prior studies examined load carriage magnitudes up to 30 % BW, the heaviest body-borne load in the present study was over 40 % BW. The lack of change in peak hip adduction may be due to the greater hip muscle activity required for load carriage running [18], as well as the increased stride frequency adopted by the participants. Load carriage running with 20 % and 30 % bodyweight has been shown to increase hip extension moment by 30 % and 60 %, respectively, compared to unloaded running [18], and this might aid in maintaining frontal plane hip stability. Furthermore, Willy et al. [33] reported that running with a 7.5 % increase in stride frequency resulted in a 2.9° reduction in peak hip adduction. It is possible that the small increases in stride frequency observed in this study (1.6%–3.6%) may have aided in counteracting any increase in hip adduction as a result of load carriage, especially in combination with the greater hip muscle activity.

The present investigation may highlight the difficulty of using variables identified as important in unloaded running to infer injury risk in a new task, which has different demands and biomechanical movement patterns. Running with a 22.7 kg body-borne load resulted in greater loading rate, free moment, and rearfoot eversion, but no change in hip adduction. So which variable should we consider as being the most important? If we only examined the free moment, we would conclude that load carriage running with 22.7 kg increases stress fracture risk; however, if we examined hip adduction, then we would conclude there is no increase in stress fracture risk. It is almost certain that load carriage running with 22.7 kg would increase stress fracture risk, as Xu et al. [18] demonstrated that load carriage running with approximately 12 and 18 kg increased tibial stress integral and the volume of the tibia experiencing higher stress magnitudes. The discrepancy in how variables do or do not change with load carriage running demonstrates that they likely do not represent the changing stress environment of the tibia, and act as poor surrogate measures for assessing stress fracture risk.

Limitations to the present study exist. All testing was performed on an instrumented treadmill, which limits the external validity of the findings. Military operations take place in an outdoor environment of variable terrain, and it is unknown how this would influence the present results. Furthermore, treadmill running is associated with a slight reduction in step length in comparison to overground running at a similar speed [34], so one might expect greater magnitudes of the variables investigated during overground running.

## 5. Conclusion

Load carriage running with 22.7 kg was associated with an increase in loading rate, peak free moment, and peak rearfoot eversion. The lack of changes in peak hip adduction may be related to the observed changes in temporospatial variables made by participants in response to the body-borne loads. Increases in stride frequency may be used to counteract the effect of load carriage, and the magnitude of the change made by participants is partially dictated by body mass. Overall, the results could indicate an elevated risk of tibial stress fracture development during heavy load carriage running; however, the lack of concomitant change amongst variables as a function of load carriage may highlight the difficulty in assessing injury risk from single external measures of running biomechanics.

## Declaration of Competing Interest

The authors declare no competing interests. The opinions and assertions contained herein are the private views of the authors and are not to be construed as official or as reflecting the views of the United States Army, the U.S. Department of Defense, or The Henry M. Jackson Foundation for the Advancement of Military Medicine, Inc. This paper has been approved for public release with unlimited distribution.

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