Effects of Stature and Load Carriage on the Running Biomechanics of Healthy Men

Junfei Tong, Adhitya V. Subramani, Vivek Kote, Michael Baggaley, W. Brent Edwards, and Jaques Reifman

***Abstract*—*Objective:* Overuse musculoskeletal injuries, often precipitated by walking or running with heavy loads, are the leading cause of lost-duty days or discharge during basic combat training (BCT) in the U.S. military. The present study investigates the impact of stature and load carriage on the running biomechanics of men during BCT. *Methods*: We collected computed tomography images and motion-capture data for 21 young, healthy men of short, medium, and tall stature (n = 7 in each group) running with no load, an 11.3-kg load, and a 22.7-kg load. We then developed individualized musculoskeletal finite-element models to determine the running biomechanics for each participant under each condition, and used a probabilistic model to estimate the risk of tibial stress fracture during a 10-week BCT regimen. *Results*: Under all load conditions, we found that the running biomechanics were not significantly different among the three stature groups. However, compared to no load, a 22.7-kg load significantly decreased the stride length, while significantly increasing the joint forces and moments at the lower extremities, as well as the tibial strain and stress-fracture risk. *Conclusion*: Load carriage but not stature significantly affected the running biomechanics of healthy men. *Significance*: We expect that the quantitative analysis reported here may help guide training regimens and reduce the risk of stress fracture.**

***Index Terms*—Finite-element modeling, individualized models, load carriage, musculoskeletal injury, tibial stress fracture.**

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I. Introduction

Musculoskeletal injuries constitute the key medical impediment to Force readiness in the U.S. military [1], accounting for more than 2.0 million medical visits annually and imposing a considerable cost on the military healthcare system [2], [3]. Moreover, overuse musculoskeletal injuries, including stress fracture (SF), are the leading cause of lost-duty days or discharge during U.S. Army basic combat training (BCT) [4]. Risk factors for musculoskeletal injuries are classified as non-modifiable (e.g., height and sex) and modifiable (e.g., physical activity and load carriage) [5]. While non-modifiable factors are helpful in identifying the population at greater injury risk, modifiable factors provide an opportunity to intervene to lower injury risk. For instance, Nindl et al. found that men and women who enter BCT with a higher physical fitness level, a modifiable risk factor, have a lower risk for developing musculoskeletal injury [6]. In particular, during strenuous physical activities, such as walking, running, and foot marching with load carriage, altering the kinematics and kinetics of the lower extremities offers an opportunity to adjust modifiable factors to help reduce injury risk [7], [8], [9], [10], [11], [12], [13], [14].

The effect of stature on musculoskeletal injury remains a controversial non-modifiable risk factor. For instance, while Beck et al. found that male Marine recruits who developed lower-extremity SF during a 12-week BCT had a significantly shorter body height and lower weight [15], Knapik et al. reported higher SF risk in taller male recruits (but not female recruits) during Army BCT [16]. In addition, Sumnik et al. found that men and women of taller stature have a higher bone mineral content and larger muscle cross-sectional area [17], which may be associated with a lower risk of SF injury [18]. Supporting this observation, our recent study involving young, healthy women running at a constant speed suggests that heavier individuals with a larger stature tend to have larger joint forces and moments, but not greater tibial strain or SF risk [7]. A history of tibial SF associated with a reduction in tibial cortical bone area has also been linked to stature. For example, Popp et al. observed a significantly smaller tibial cortical area in a group of shorter women with a SF history, when compared with a group of taller women without an injury history [19]. In contrast, Cosman et al. reported that, for the same stature, in men (but not in women), tibial cortical bone area is smaller in those with a history of tibial SF versus those without a previous injury [20]. Additional

TABLE I
ANTHROPOMETRIC CHARACTERISTICS OF 21 YOUNG, HEALTHY MEN

Stature group	Age (year)	Mass (kg)	Height (m)	Foot length (m)	Body fat (%)	BMI (kg/m^2)
Short (N=7)	19.4 (1.4)	69.0 (4.7)	1.71 (0.05)	0.27 (0.01)	8.5 (1.7)	23.8 (1.9)
Range	18-21	63.0-76.0	1.62-1.74	0.26-0.28	6.5-11.5	20.8-26.3
Medium (N=7)	19.6 (1.0)	71.3 (6.8)	1.76 (0.02)	0.27 (0.01)	8.6 (1.9)	22.9 (2.2)
Range	18-21	60.0-77.5	1.75-1.79	0.26-0.28	6.2-11.6	19.3-25.6
Tall (N=7)	19.7 (1.3)	75.5 (6.4)	1.84 (0.02)	0.28 (0.01)	9.1 (2.4)	22.3 (1.8)
Range	18-21	68.0-83.7	1.82-1.88	0.26-0.29	6.0-13.2	20.1-25.0
***p* value**	0.909	0.149	**<0.001**	0.097	0.846	0.402

The data are presented as means (1 standard deviation) or range. A bold *p* value indicates the parameter is significantly different at the 0.05 level among the three stature groups, based on an analysis of variance test. BMI: body mass index.

work is needed to elucidate the impact of stature on the running biomechanics of men.

Independent of stature, multiple studies have found that load carriage substantially affects the biomechanics of the lower extremities [7], [8], [9], [10], [11], [12], [13], [14]. In general, load carriage while walking or running increases ground reaction forces (GRFs) and joint reaction forces (JRFs), while increasing ground-contact time and decreasing stride length [11]. The increased loading of the lower-extremity joints (e.g., knee and ankle) not only contributes to joint injuries [11], but also increases tibial stress and strain, contributing to tibial SF injuries observed in Army personnel [21], [22]. For example, previous studies involving women walking or running while carrying loads of 20-23 kg found significant increases in peak tibial stress and strain [7], [21], [22], which could lead to an increase in tibial SF risk [7]. Without load carriage, a shortened stride length associated with alterations in lower-extremity joint kinematics (i.e., hip and knee flexion) [23] reduces the risk of SF injuries [24], [25]. However, the impact of load carriage on the joint kinematics of the lower extremities is less conclusive. For example, while Loverro et al. [10] and Silder et al. [26] reported an increase in hip and knee flexion in men and women while walking or running with a load, Brown et al. found no significant changes in lower-extremity joint kinematics in men running with a load [12]. In our recent study of young, healthy women, we found that running with a load significantly increases hip flexion, JRFs, tibial strain, and tibial SF risk [7]. However, we do not know the extent to which carrying a load would impact the running biomechanics of men, nor do we know whether the impact of load depends on stature. The answers to these research questions could have practical ramifications if, in an attempt to reduce musculoskeletal injuries during BCT, the U.S. Army decided to adjust load carriage based on body stature. The Army recruits men with a wide range of body statures [27], however, currently, each recruit regardless of height is required to carry the same load (i.e., 0 to 23 kg) [28].

The objective of this study is to quantify the effects of stature and load carriage on the running biomechanics of young, healthy men, including stride spatiotemporal parameters, kinematics and kinetics of the lower-extremity joints, tibial strain, and tibial SF risk. Towards this end, we developed *individualized* musculoskeletal finite-element (FE) models based on newly collected experimental data for 21 men of three statures (short, medium, and tall), while running with a 0-kg, 11.3-kg, and 22.7-kg load. We hypothesized that the running biomechanics of young, healthy men are dependent on both stature and load carriage. Specifically, we hypothesized that taller men would have larger joint forces and moments, greater tibial strain, and higher tibial SF risk. In addition, for all groups, we hypothesized that an external load carriage would increase stance duration, joint angles, forces and moments, as well as tibial strain and SF risk.

II. METHODS

A. Image Acquisition and Motion-Capture Data Collection

We enrolled 21 young, healthy men of three statures (short, medium, and tall) between the ages of 18 and 21 years, as representative of military recruits. We set the height criteria as less than the 45th percentile (<1.75 m; short), between the 45th percentile and 70th percentile (1.75 m to 1.79 m; medium), and greater than the 80th percentile (>1.81 m; tall) of the U.S. male Soldier population [27]. We established such recruitment criteria in an attempt to enlist participants for three distinct stature groups that approximately represented each tertile of Army personnel, while trying to create gaps between the groups. All participants were self-reported experienced treadmill runners and free from injuries limiting their physical activity 3 months prior to enrollment in the study. For each participant, we recorded their age, mass, height, foot length, body fat percentage, and body mass index (BMI), and averaged the values within each stature group (Table I). We also collected quantitative computed tomography (CT) images of the left tibia of each participant using a GE Discovery Scanner (General Electric Medical System, Milwaukee, WI). The scans had an in-plane pixel resolution of 0.49×0.49 mm^2 and a slice thickness of 0.63 mm. Each CT scan included a calibration phantom of known calcium hydroxyapatite concentration (QRM, Moehrendorf, Germany) in the field of view.

Each participant completed three running trials in a randomized order at a constant speed of 3.0 m/s on an instrumented treadmill (Bertec Corporation, Columbus, OH), including trials with no load, an 11.3-kg load (25 lb), and a 22.7-kg load (50 lb). We selected this moderate running speed and the three load conditions as representative values of U.S. Army BCT [28]. The load was symmetrically distributed over the front and back because during BCT military personnel carry approximately symmetrical loads >90% of the time [29], and we adjusted the load using a V-max vest (V-max, Rexburg, ID). Similar to our prior studies [7], [30], we collected motion-capture data at 200 Hz using an eight-camera motion-analysis system (Vicon Nexus, Centennial, CO). We placed 42 retroreflective markers bilaterally on anatomical landmarks and segments, including the arm, trunk, pelvis, thigh, shank, and foot, and synchronously collected force-platform data at 1000 Hz. For each trial, we collected 20 seconds of data after each participant had reached a steady-state stride of 3.0 m/s. This length of data provided a sufficient number of strides (>20) to obtain consistent stance and swing durations for all conditions from which to select a representative stride.

The study protocol was approved by the University of Calgary Conjoint Health Research Ethics Board and by the Human Research Protection Office at the U.S. Army Medical Research and Development Command, Fort Detrick, MD. A written informed consent was obtained from each participant before enrollment in the study.

B. Stride Analysis and Individualized Musculoskeletal Model

Prior to performing individualized musculoskeletal analysis for each participant, we first selected one representative stride from the 20-second recordings, because we cannot aggregate the motion-capture recordings and use their average or median values for the computational models. To select a representative stride for each participant in a trial, we followed the procedure proposed by Sangeux and Polak [31], which consists of the following five steps: *1*) identify the start and end points of each stride (and, hence, the stride duration) for the 20-second data by setting a threshold of 25 N for the vertical GRF; *2*) re-sample the GRF time history of each stride so that all strides of different durations are represented by 100 GRF values; *3*) determine the median GRF time history stride based on all re-sampled strides in the trial; *4*) compute the distance from each re-sampled stride to the median stride; and *5*) select the one closest to the median GRF time history stride as the representative stride for a given trial. We then computed the length of the representative stride by multiplying the duration and the running speed (3.0 m/s). Finally, we normalized the stride length by dividing it by the participant's height.

Similar to our prior studies [7], [13], [14], to determine the joint kinematics and kinetics for each participant, we developed individualized musculoskeletal models using the AnyBody Modeling System (AnyBody Technology, Aalborg, Denmark). Briefly, the software provides a generic musculoskeletal model of a male subject consisting of rigid segments, including arms, trunk, pelvis, thighs, shanks, and feet, as well as 55 muscles for each leg. So, to develop an individualized musculoskeletal model, we first morphed the generic tibial geometry in the AnyBody model to match the individualized tibial geometry extracted from the CT scan using Mimics (Materialise, Leuven, Belgium). Specifically, we used the automated segmentation built in Mimics to segment the bone and tissues, followed by a manual refinement to generate a clean tibial geometry. Next, we scaled the other segments in the generic musculoskeletal model based on the anthropometric measures (e.g., mass, height, foot length, and body fat percentage). Then, we applied an optimization scheme that minimizes the errors between markers tracked in the experiment and markers defined in the model, to further optimize the segments' length. Once optimized, using the marker-tracking data for the representative stride, we computed the body motion (i.e., the joint angular changes throughout the entire body), including the kinematics of the lower-extremity joints (e.g., hip, knee, and ankle). Finally, we determined the kinetics of the hip, knee, and ankle by performing an inverse dynamic analysis and normalizing the GRFs and JRFs by body weight (BW) and the joint moments by body mass, for each of the three running conditions for each participant. As often reported [32], here we provided GRFs in the vertical direction relative to the ground because it had by far the highest magnitude. For the JRFs, for consistency, we reported the resultant forces reflecting the three directions because for the hip and ankle the largest contributor was in one direction (proximal-distal) and for the knee it was in two directions (proximal-distal and anterior-posterior). For the joint moments, we reported them in the sagittal plane because moments in different directions cannot be added and the highest joint moment occurred predominantly in the sagittal plane.

C. Individualized Finite-Element Analysis

We performed the individualized FE analysis using the same steps as in our previous work [13], except for the host-mesh fitting method, because here we collected CT images for each subject. Briefly, we created the three-dimensional (3-D) FE model of the tibia using subject-specific tibial geometry extracted from the CT scan. Then, using HyperMesh software (Altair Engineering, Inc., Troy, MI), we meshed the 3-D FE model using 10-noded quadratic tetrahedral elements, with an average element size of 3.0-3.5 mm. We assumed that all elements were linear elastic and isotropic, with heterogeneous Young's moduli, where we determined the Young's modulus of each element based on the Hounsfield units of the CT scan. In addition, we identified the elements with Young's moduli greater than 8 GPa as cortical bone, between 8 GPa and 6 MPa as trabecular bone, and the remaining elements as intramedullary tissue [33]. As the bone and tissue components have different Poisson's ratios, we assigned a ratio of 0.325 to the bone elements and 0.167 to the intramedullary tissue elements [34].

We determined the loading conditions for the FE model based on the muscle forces, joint forces, and joint moments from the inverse dynamic analysis. Specifically, we coupled the muscle and ligament insertion points in the individualized musculoskeletal model with the outer surface of the tibial mesh as FE constraint nodes. In total, we created 171 couplings for each FE model. We then performed FE analysis to determine the tibial strain for each element using Abaqus 2019 (Dassault Systèmes, Vélizy-Villacoublay, France). We calculated the von Mises strain for each cortical bone element and determined the peak von Mises strain (90th percentile) of the tibial cortical bone for the representative stride.

D. Probabilistic Model for Predicting Stress-Fracture Risk

We predicted tibial SF risk using a probabilistic model based on bone fatigue damage, while accounting for bone repair and adaptation, and making slight modifications to our previous work [7], [35]. Briefly, to account for the changes in tibial strain due to bone adaptation, we calculated the equivalent strain $\Delta\varepsilon_{\mathrm{eq}}$ based on the von Mises strain $\Delta\varepsilon$ of each cortical element, according to

$$\Delta\varepsilon_{eq} = \left[\frac{1}{t_d}\int_0^{t_d}(R * \Delta\varepsilon)^n dt\right]^{1/n} \quad (1)$$

where t_d denotes the duration of bone adaptation in days, R represents the strain adaptation ratio [36], and n denotes a material constant, which we set to 6.6 [35]. In this process, we excluded the top 1% of the elements exhibiting extreme strain values because such values, likely caused by numerical artifacts in the FE analysis (i.e., singularities caused by point loads in muscles and ligaments), may not be physiological. Then, we divided the tibia into 16 groups, with each group experiencing similar strain levels, and computed the group-wise strain ε_i, with $i = 1, 2, \ldots, 16$, by averaging $\Delta\varepsilon_{eq}$ within all elements of the ith group. Using this information, we estimated the fatigue life t_{fi} in days of the ith group, according to

$$t_{fi} = \frac{C * \varepsilon_i^{-n}}{N_L} \quad (2)$$

where C (set to 2.1×10^{32}) denotes the fatigue constant derived from a human tibia beam-bending experiment [37] and N_L represents the number of loading cycles per day. Given activity duration t_a in days, we determined the SF risk P_{fi} of the ith group of the tibia, as follows

$$P_{fi} = 1 - \exp\left[-\frac{V_i}{V_o} * \left(\frac{t_a}{t_{fi}}\right)^{1.2}\right] \quad (3)$$

where V_o set to 96 mm^3 denotes the volume of the cortical bone sample in the experiment (i.e., a normalization factor) and V_i represents the sum of the volumes of all the elements in the ith group.

Finally, we determined the tibial SF risk P_s, which incorporates bone repair and adaptation, as follows

$$P_s = \int_0^{t_d} Q_f * (1 - P_r)\, dt \quad (4)$$

where Q_f denotes the time differential of the entire tibia SF risk P_f for both legs and P_r denotes the rate of bone repair. Here, P_f was estimated as follows

$$P_f = 1 - \prod_{i=1}^{16}(1 - P_{fi})^2 \quad (5)$$

and P_r was modeled through a Weibull equation as follows

$$P_r = 1 - \exp\left[-\left(\frac{t_d}{26}\right)^{2.0}\right] \quad (6)$$

where 26 denotes the reference time for the bone-repair process (i.e., a normalization factor) and 2.0 denotes the Weibull modulus [35].

E. Individualized Risk Prediction for a 10-Week BCT

To assess the impact of load carriage during a typical 10-week BCT [28], we utilized the probabilistic model to predict the tibial SF risk for each participant in the study while running with and without a load. Towards this end, we estimated the daily duration of strenuous physical activities during the BCT that were expected to affect the risk of tibial SF. Accordingly, after review of the large set of activities described in the Holistic Health and Fitness publication, which establishes the Army's training and testing doctrine for achieving Soldier readiness for 21st century warfare [28], we only considered running and foot marching. Over a 10-week BCT, we estimated that recruits run for a total of 453 minutes and foot march for 800 minutes. From a SF-risk perspective, we then estimated the running time equivalent to 800 minutes of foot marching, which added only 5.6 minutes to a "total" running time of 458.6 minutes. We estimated the additional running time through the following steps: *1*) based on our prior study [21], we assumed that the tibial strain during walking was only one-half that of running; *2*) using Eq. (2), we determined that the cumulative fatigue damage from 100 walking strides was equivalent to one running stride; and *3*) assuming that the marching frequency was ~60 strides/minute and that the running frequency was ~85 strides/minute, we determined that 800 minutes of foot marching was equivalent to 5.6 minutes [(800*60)/(85*100)] of running. Then, we defined a representative week, which was repeated for 10 consecutive weeks, where for each week we assumed the following running schedule: no running for the third and seventh days, and running 1.7 km/day (at a 3.0 m/s constant speed) for the remaining five days. For each participant, we estimated the number of loading cycles per day by dividing the daily running distance by the participant's representative load-condition-dependent stride length.

F. Statistical Analysis

Prior to participant recruitment, we performed a power analysis and determined that 7 subjects per group were sufficient to provide group-based differences. We calculated the sample size using group means and standard deviations (SD) of the peak vertical GRF and leg stiffness for individuals who ran while carrying a load [26]. Based on the calculated effect size (0.75), we determined that a sample size of 21 was sufficient to observe a statistically significant difference ($p < 0.05$) with a statistical power of 0.80.

For the anthropometric characteristics, we performed analysis of variance to identify statistically significant differences among the three stature groups. To determine the impact of stature and load carriage on the running biomechanical responses, we developed linear mixed-effects models for various dependent variables, including spatiotemporal parameters (e.g., stride duration and normalized stride length), joint kinematics, joint kinetics, peak tibial strain, and tibial SF risk. We treated stature and load as fixed categorical effects (interaction of stature and load also included), and the subject as a random effect (random intercept only). For each dependent variable, listed in the first column of Tables II and III, we first evaluated the significance of the interaction term in the model using the Wald *F*-test with the Kenward-Roger approximation for degrees of freedom [38]. When the interaction term was not statistically significant, we removed it from the model, developed a new linear mixed-effects model with only two main effects, and evaluated the statistical significance of each effect using the Wald *F*-test. If only one of the effects was statistically significant, we grouped the data for the non-significant term and performed pairwise analysis for the significant effect, using post hoc Tukey's pairwise comparisons

TABLE II
SPATIOTEMPORAL PARAMETERS AND PEAK JOINT ANGLES

	Short (N=7)			Medium (N=7)			Tall (N=7)			*p* value	
Load (kg)	**0.0**	**11.3**	**22.7**	**0.0**	**11.3**	**22.7**	**0.0**	**11.3**	**22.7**	**Load**	**Stature**
Normalized stride length											
	1.26 (0.10)	1.24 (0.09)	1.22 (0.09)	1.21 (0.05)	1.19 (0.06)	1.17 (0.06)	1.24 (0.05)	1.25 (0.06)	1.22 (0.05)	**<0.001**	0.363
Stance duration (s)											
	0.26 (0.03)	0.28 (0.02)	0.30 (0.03)	0.26 (0.01)	0.28 (0.01)	0.29 (0.01)	0.28 (0.03)	0.30 (0.03)	0.31 (0.02)	**<0.001**	0.267
Peak joint angle (degrees)											
Hip											
Flex	41.0 (8.0)	40.3 (7.7)	40.1 (9.0)	35.9 (8.5)	33.8 (4.9)	35.2 (5.1)	34.3 (3.3)	35.7 (3.2)	33.8 (3.2)	0.713	0.139
Ext	17.3 (9.3)	18.6 (9.9)	18.4 (9.5)	22.8 (6.2)	23.1 (6.4)	22.8 (5.7)	25.3 (6.2)	25.2 (5.8)	26.0 (6.4)	0.580	0.184
Knee											
Flex	46.6 (8.0)	46.3 (7.3)	47.4 (8.4)	45.8 (3.2)	45.6 (3.1)	45.0 (3.3)	45.3 (4.0)	47.2 (4.5)	45.6 (4.0)	0.555	0.901
Ankle											
DF	37.5 (5.3)	38.8 (6.1)	40.3 (5.8)	36.0 (4.9)	37.7 (5.8)	37.9 (6.2)	36.3 (4.8)	38.1 (4.7)	38.8 (4.3)	**<0.001**	0.831
PF	17.7 (8.0)	18.4 (8.6)	17.3 (9.5)	15.5 (8.0)	17.0 (6.5)	17.2 (5.8)	19.4 (8.9)	20.2 (8.7)	18.7 (8.2)	0.404	0.787

The data are averaged within the group of short, medium, and tall men, and presented as means (1 standard deviation). Bold *p* indicates statistically significant main effect (load or stature) based on a mixed-effects model. DF: dorsiflexion; Ext: extension; Flex: flexion; PF: plantarflexion.

TABLE III
PEAK GROUND REACTION FORCES, JOINT KINETICS, TIBIAL STRAIN, AND TIBIAL STRESS-FRACTURE RISK

	Short (N=7)			Medium (N=7)			Tall (N=7)			*p* value	
Load (kg)	**0.0**	**11.3**	**22.7**	**0.0**	**11.3**	**22.7**	**0.0**	**11.3**	**22.7**	**Load**	**Stature**
Ground reaction force (BW)											
	2.5 (0.3)	2.7 (0.3)	2.8 (0.4)	2.4 (0.1)	2.6 (0.1)	2.8 (0.2)	2.5 (0.2)	2.7 (0.2)	2.9 (0.2)	**<0.001**	0.698
Joint reaction forces (BW)											
Hip	9.0 (2.0)	9.4 (2.0)	10.1 (2.1)	7.3 (0.9)	7.9 (1.0)	8.5 (1.0)	7.8 (0.8)	8.5 (0.5)	9.3 (1.1)	**<0.001**	0.107
Knee	11.6 (1.4)	12.2 (1.4)	12.8 (1.4)	11.3 (1.2)	12.3 (1.6)	12.8 (1.9)	11.8 (1.1)	13.0 (1.1)	13.8 (1.0)	**<0.001**	0.535
Ankle	11.2 (2.2)	11.8 (1.6)	12.4 (1.5)	11.9 (2.2)	12.7 (2.6)	13.4 (2.9)	11.9 (1.5)	12.7 (1.3)	13.6 (1.2)	**<0.001**	0.586
Joint moments (N•m/kg)											
Hip											
Flex	0.9 (0.2)	1.0 (0.2)	1.1 (0.2)	1.0 (0.1)	1.0 (0.1)	1.1 (0.2)	0.9 (0.1)	0.9 (0.1)	1.1 (0.2)	**<0.001**	0.626
Ext	2.4 (0.5)	2.8 (0.3)	3.0 (0.4)	2.0 (0.6)	2.2 (0.6)	2.5 (0.4)	2.3 (0.3)	2.6 (0.5)	2.7 (0.4)	**<0.001**	0.113
Knee Ext	2.2 (0.6)	2.3 (0.6)	2.5 (0.6)	2.0 (0.5)	2.2 (0.6)	2.3 (0.7)	2.3 (0.2)	2.6 (0.3)	2.6 (0.3)	**<0.001**	0.509
Ankle PF	2.9 (0.4)	2.9 (0.2)	3.0 (0.3)	3.1 (0.5)	3.2 (0.6)	3.4 (0.7)	3.2 (0.4)	3.3 (0.4)	3.5 (0.4)	**<0.001**	0.270
Tibial strain (με)											
	4,362 (817)	4,591 (975)	4,801 (1,054)	4,648 (1,256)	4,998 (1,509)	5,184 (1,787)	4,610 (662)	4,915 (550)	5,122 (572)	**<0.001**	0.799
Tibial SF risk (%)											
	0.7 (3.6)	1.1 (3.9)	1.6 (4.2)	1.1 (7.2)	1.8 (6.8)	2.1 (9.4)	1.5 (4.0)	2.6 (2.9)	3.5 (2.9)	**<0.001**	0.625

The data are averaged within the group of short, medium, and tall men, and presented as means (1 standard deviation). Bold *p* indicates statistically significant main effect (load or stature) based on a mixed-effect model. BW: body weight; Ext: extension; Flex: flexion; PF: plantarflexion; SF: stress fracture.

with Holm-Bonferroni correction [39]. In addition, we also computed the effect size (Cohen's d) [40] between the short and tall groups for dependent variables for which stature had a significant effect, and between the no-load and 22.7-kg-load conditions for dependent variables for which load had a significant effect. For SF risk, which was not normally distributed, we performed a log transformation prior to the analysis. All data are presented as means (SD), unless otherwise noted. We performed all statistical analyses using the RStudio v1.4 statistical software, including the *lme4*, *lmerTest, and emmeans* packages with an alpha level of 0.05.

III. Results

A. Spatiotemporal Parameters and Joint Kinematics

Table II shows the mean (SD) values of the spatiotemporal parameters and joint kinematic parameters calculated using the musculoskeletal models. We found that none of these parameters was significantly different among the three stature groups. In contrast, the normalized stride length, stance duration, and peak ankle dorsiflexion were significantly different among the three loading conditions (Tables II and IV and Fig. 1). For example, when compared to the baseline no-load condition, a load of 22.7 kg significantly decreased the normalized stride length by 1.6% [1.23 (0.07) to 1.21 (0.07), effect size = 0.7] (Fig. 1(a)). Compared to the baseline condition, a load of 11.3 kg significantly increased the stance duration by 11.5% [0.26 s (0.02) to 0.29 s (0.02)] (Fig. 1(b)). We observed an even larger increase for the 22.7-kg load (i.e., 15.4%, effect size = 2.7). Finally, a load of 22.7 kg significantly increased the peak ankle dorsiflexion by 6.6% [36.6 (4.8) degrees to 39.0 (5.3) degrees, effect size = 0.9] (Fig. 1(c)).

B. Ground Reaction Force and Joint Kinetics

We found that the GRF and joint kinetic parameters were only significantly different among the three loading conditions (Tables III and IV). As expected, compared to the baseline no-load condition, running with a load of 11.3 kg or 22.7 kg

TABLE IV
POST HOC PAIRWISE ANALYSIS

	Load (kg)			p value for post hoc analysis*		
	0.0	11.3	22.7	0 kg vs 11.3 kg	11.3 kg vs 22.7 kg	0 kg vs 22.7 kg
Normalized stride length	1.23 (0.07)	1.23 (0.07)	1.21 (0.07)	0.426	**0.014**	**<0.001**
Stance duration (s)	0.26 (0.02)	0.29 (0.02)	0.30 (0.02)	**<0.001**	**<0.001**	**<0.001**
Peak ankle dorsiflexion (degrees)	36.6 (4.8)	38.2 (5.3)	39.0 (5.3)	**0.003**	0.213	**<0.001**
Peak ground reaction force (BW)	2.5 (0.2)	2.6 (0.2)	2.8 (0.3)	**<0.001**	**<0.001**	**<0.001**
Peak joint reaction force (BW)						
Hip	8.0 (1.5)	8.6 (1.4)	9.3 (1.5)	**<0.001**	**<0.001**	**<0.001**
Knee	11.5 (1.2)	12.5 (1.4)	13.1 (1.5)	**<0.001**	**0.001**	**<0.001**
Ankle	11.7 (1.9)	12.4 (1.8)	13.1 (2.0)	**0.002**	**<0.001**	**<0.001**
Peak joint moments (N•m/kg)						
Hip Flex	0.9 (0.1)	1.0 (0.1)	1.1 (0.2)	0.234	**0.007**	**<0.001**
Hip Ext	2.2 (0.5)	2.5 (0.5)	2.7 (0.4)	**<0.001**	**0.034**	**<0.001**
Knee Ext	2.2 (0.4)	2.3 (0.5)	2.5 (0.5)	**0.003**	**0.020**	**<0.001**
Ankle PF	3.0 (0.5)	3.1 (0.5)	3.3 (0.5)	**0.049**	**0.003**	**<0.001**
Tibial strain (με)	4,540 (906)	4,835 (1,045)	5,036 (1,191)	**0.010**	0.102	**<0.001**
Tibial SF risk (%)	1.1 (4.6)	1.7 (4.3)	2.3 (5.0)	**0.008**	0.123	**<0.001**

The data are averaged for all 21 men and presented as means (1 standard deviation). Bold *p* indicates statistically significant differences between two load conditions based on post hoc analysis. **p* values adjusted for multiple comparisons using the Holm-Bonferroni correction. BW: body weight; Ext: extension; Flex: flexion; PF: plantarflexion; SF: stress fracture.

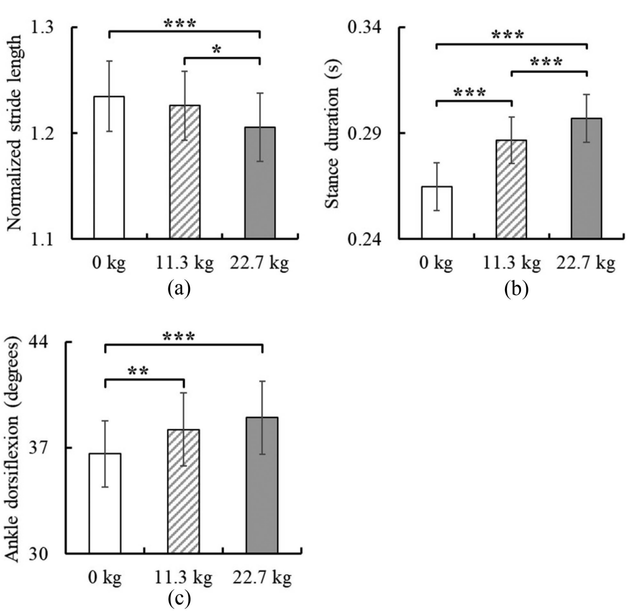

Fig. 1. Comparison of (a) Normalized stride length, (b) Stance duration, and (c) Peak ankle dorsiflexion while running with no load, an 11.3-kg load, and a 22.7-kg load. N = 21 under each load condition. Error bar: 95% confidence interval. $*p < 0.05$, $**p < 0.01$, $***p < 0.001$, based on post hoc Tukey's pairwise comparisons with Holm-Bonferroni correction.

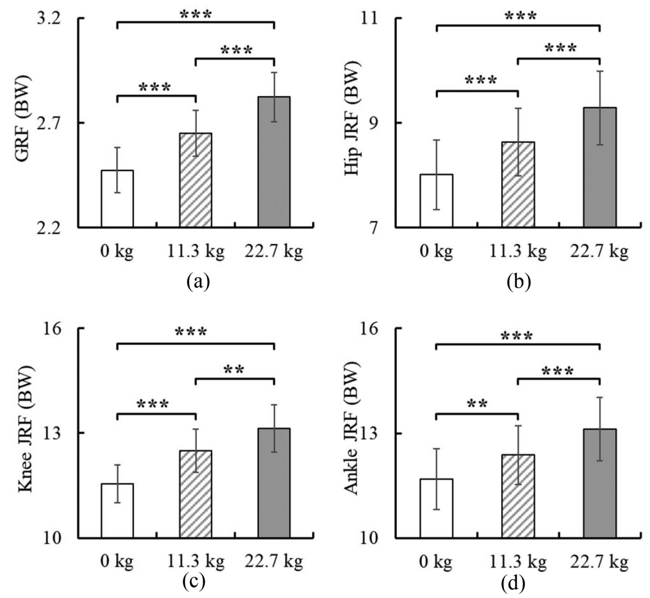

Fig. 2. Comparison of (a) Normalized peak ground reaction force (GRF) and (b-d) peak joint reaction force (JRF) at the hip, knee, and ankle while running with no load, an 11.3-kg load, and a 22.7-kg load. N = 21 under each load condition. BW: Body weight. Error bar: 95% confidence interval. $*p < 0.05$, $**p < 0.01$, $***p < 0.001$, based on post hoc Tukey's pairwise comparisons with Holm-Bonferroni correction.

significantly increased the peak GRF by 4.0% or 12.0% (effect size = 2.5), respectively (Fig. 2(a)). Compared to the baseline condition, a load of 22.7 kg significantly increased the peak JRF at the hip, knee, and ankle (Fig. 2(b)–(d)). For example, running with a 22.7-kg load increased the peak hip JRF by 16.3% [8.0 (1.5) BW to 9.3 (1.5) BW, effect size = 1.9], increased the peak knee JRF by 13.9% [11.5 (1.2) BW to 13.1 (1.5) BW, effect size = 1.9], and increased the peak ankle JRF by 12.0% [11.7 (1.9) BW to 13.1 (2.0) BW, effect size = 1.6].

Consistent with the JRF results, we found that the peak joint moments at the hip, knee, and ankle also significantly increased when running with a 22.7-kg load (Fig. 3). For instance, compared to the baseline no-load condition, the peak hip flexion moment increased by 22.2% [0.9 (0.1) N•m/kg to 1.1 (0.2) N•m/kg, effect size = 0.8]. Similarly, the peak hip extension moment increased by 22.7% [2.2 (0.5) N•m/kg to 2.7 (0.4) N•m/kg, effect size = 1.4]. Compared to the baseline condition, the heavier load significantly increased the peak knee extension moment by 13.6% [2.2 (0.4) N•m/kg to 2.5 (0.5) N•m/kg, effect size = 1.1]. Finally, we found that running with the heavier load significantly increased the peak ankle plantarflexion moment by 10.0% [3.0 (0.5) N•m/kg to 3.3 (0.5) N•m/kg, effect size = 1.2].

C. Tibial Strain and Stress-Fracture Risk

We found that the peak von Mises strain was located at the posteromedial cortex of the tibial shaft for more than two-thirds of the participants. For a given load, we did not observe significant differences in peak tibial strain among the three stature groups. In contrast, compared to the baseline condition, we found that running with a load of 11.3 kg or 22.7 kg significantly increased the peak tibial strain (Table IV and Fig. 4(a)). Specifically, a

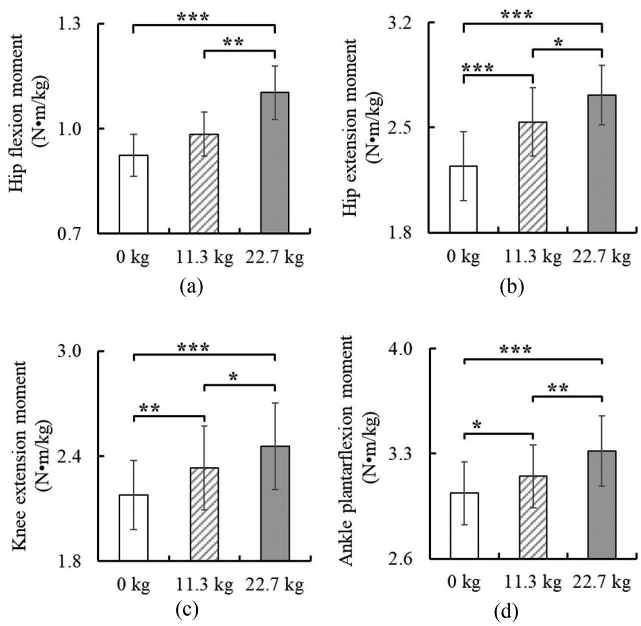

Fig. 3. Comparison of normalized peak joint moments at the (a-b) Hip, (c) Knee, and (d) Ankle while running with no load, an 11.3-kg load, and a 22.7-kg load. N = 21 under each load condition. Error bar: 95% confidence interval. $*p < 0.05$, $**p < 0.01$, $***p < 0.001$, based on post hoc Tukey's pairwise comparisons with Holm-Bonferroni correction.

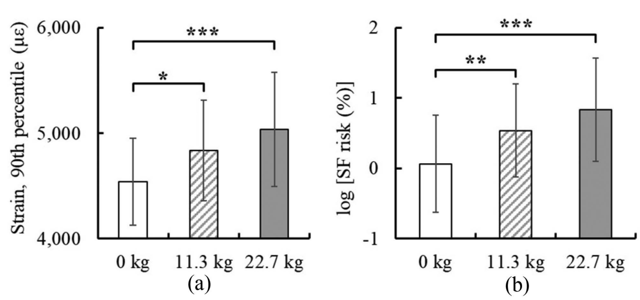

Fig. 4. Comparison of (a) Von Mises strain (90th percentile) of the tibia and (b) Log-transformed stress fracture (SF) risk while running with no load, an 11.3-kg load, and a 22.7-kg load. N = 21 under each load condition. Error bar: 95% confidence interval. $*p < 0.05$, $**p < 0.01$, $***p < 0.001$, based on post hoc Tukey's pairwise comparisons with Holm-Bonferroni correction.

load of 11.3 kg increased the peak tibial strain by 6.5% [4540 (906) $\mu\varepsilon$ to 4835 (1045) $\mu\varepsilon$] and a load of 22.7 kg increased the peak tibial strain by 10.9% [4540 (906) $\mu\varepsilon$ to 5036 (1191) $\mu\varepsilon$, effect size = 1.0]. Similarly, we did not observe significant differences in the predicted tibial SF risk among the three stature groups. However, compared to the baseline condition, running with a load of 11.3 kg or 22.7 kg significantly increased the SF risk (Table IV and Fig. 4(b)). For example, a load of 11.3 kg increased the SF risk by 54.5% [1.1% to 1.7%] and a load of 22.7 kg increased the SF risk by 109.1% [1.1% to 2.3%, effect size = 1.0].

IV. Discussion

With the goal of analyzing the effects of stature and load carriage on the running biomechanics of young, healthy men, we collected experimental data for 21 men of short (N = 7), medium (N = 7), and tall (N = 7) statures, developed individualized musculoskeletal and FE models, and predicted the risk of tibial SF for each participant for a running regimen representative of a 10-week BCT in the U.S. Army. In particular, we assessed the effects of stature and load carriage on stride duration, normalized stride length, joint kinematics, joint kinetics, peak tibial strain, and tibial SF risk. Contrary to our hypothesis, we found that none of these variables was significantly different among the three stature groups. In partial agreement with our hypothesis, when compared to the baseline no-load condition, running with a load, in particular the heavier load (22.7 kg), significantly affected the running biomechanics of the participants (Figs. 1–4 and Tables II–IV).

We found that load carriage significantly affected the running biomechanics of young, healthy men, similar to our prior study involving young, healthy women of different statures [7]. Compared to baseline, the normalized stride length decreased slightly (1.6%), but significantly, when running with a 22.7-kg load (Fig. 1(a)). In contrast, compared to baseline, stance duration increased significantly by 11.5% when running with an 11.3-kg load and by 15.4% with a 22.7-kg load (Fig. 1(b)). These results are consistent with the work of Willy et al., which reported that running at a fixed speed (2.7 m/s) with a 15-kg load decreases stride length by 2.2% and increases stance duration by 6.3% [11]. The longer stance duration when running with a load may indicate an increase in metabolic cost [41]. In addition, a prior study showed that manipulating stride length, such as a 10% decrease, reduces tibial SF risk by 3% to 6% when running at a self-preferred speed [25]. Hughes et al. [22] found that carrying a 30-kg load significantly increases the tibial compressive strain by more than 10%, which could lead to an increase in tibial SF risk. In our study, the stride length reduction when running with a load was minor (less than 2%), and its impact on SF risk was counteracted by the load [30].

Compared to baseline, we did not find significant changes in peak knee or hip flexion in men running with a load (Table II). These findings contradict the work of Silder et al. [26], who observed a significant increase in knee and hip flexion in their combined analysis of women and men running at a self-preferred speed with a 30% BW load. However, our findings are consistent with the work of Brown et al. [12], who found no change in knee or hip flexion in men running at a fixed speed (3.5 m/s) with a 40-kg load. This contradiction may be explained by the fact that Silder et al. analyzed men and women together as one group, and the observed differences may have been present in women but not men when running with an external load. In fact, in our prior study of women running with a load [7], we did observe a significant increase in hip flexion with a 22.7-kg load. In addition, compared to baseline, we also observed an increase of 6.6% in peak ankle dorsiflexion while running with a 22.7-kg load (Fig. 1(c)). Rice et al. suggested that such an increase of ankle dorsiflexion lowers the body center of mass so as to maintain postural stability [9].

Similar to our prior study in women [7], compared to baseline, GRFs, JRFs, and joint moments significantly increased when running with a load, especially a heavy load (Figs. 2 and 3). We observed the highest increase in JRFs at the hip, followed by the knee and then the ankle, which was consistent with our prior study [14]. Such an increase in JRFs may lead to an increased risk of musculoskeletal injuries at the lower extremities [42]. In addition, we observed a significant increase in joint moments at the hip, knee, and ankle (Fig. 3). We speculate that the increase in moments indicates that additional muscle forces are required to generate the necessary moments when running with an extra load, possibly leading to an increase in metabolic costs and earlier muscle fatigue [43].

Supporting our hypothesis, compared to baseline, running with an 11.3-kg or a 22.7-kg load significantly increased the peak tibial strain (Fig. 4(a)), including a 10.9% increase for the heavy load (22.7 kg). Reassuringly, we found that the location of the peak strain (posteromedial cortex of the tibial shaft) coincided with the injury location described in a previous publication [44]. Using these strain data, we predicted tibial SF risk during BCT, which increased with load (Fig. 4(b)). Similar to strain, the increase was statistically significant for running with an 11.3-kg or a 22.7-kg load, with the heavier load increasing the risk by more than 100% compared to the no-load condition. These results support existing Army guidelines that discourage running with a load [45]. Overall, across stature groups and load conditions, the tibial SF risk during a 10-week BCT ranged from 0.7% to 3.5%, which is consistent with the reported incidence rate of 0.8% to 5.0% of SF in the tibia for young, healthy men during BCT [46]. We note that an outlier in the medium group caused the tibial strain in that group to be higher than that of the tall group (Table III), which is not reflected in the SF risk because, prior to the linear mixed-effects model analysis, we applied a log transformation to the data.

In addition, while the differences in SF risk among the three stature groups were not statistically significant, which contradicted our hypothesis, we observed a trend of increasing risk with taller statures, yielding a medium effect size of 0.5 between the short and tall groups (Table III). The work by Beck et al. contradicts our findings, as they found that the SF group has shorter body height and lower weight [15]. However, the study by Knapik et al. [16], which consisted of nearly 500000 male participants (vs. 626 for Beck et al.), showed that taller or heavier men have a higher SF risk, supporting our observations.

To investigate the impact of extended running during BCT, we doubled the running distance from 1.7 km/day to 3.4 km/day five days a week for each of the 10 weeks of BCT, and re-estimated the SF risk. Doubling the running distance at a constant speed of 3.0 m/s increased the SF-injury risk by $\sim$120% for both the no-load and 22.7-kg load conditions.

Our study has several limitations. First, the running experiment was performed at a self-preferred stride length at a fixed speed (3.0 m/s) on a level treadmill. Hence, to maintain the fixed speed, participants in the short group had higher stride frequency than those in the tall group to account for their shorter stride length, and the differences in frequency varied slightly for the three load conditions (from 6.6% to 8.5%). In turn, the higher stride frequency could lead to a lower peak knee flexion and peak hip flexion [23], and it could also reduce the peak vertical GRF and lower limb joint kinetics (e.g., knee and hip extension moments) [47]. Alternatively, the short group could have kept pace by increasing their stride length. Such "overstride," which did not actually occur, would have increased the stress in the tibia due to the greater moments (i.e., knee extension and ankle plantarflexion moments) at the two ends of the tibia [47], increasing the estimated SF-injury risk in this group and causing the differences in risk between the three groups to be even smaller than the ones reported here. In addition, the conclusions may not be applicable to running at a varied speed or running on a graded treadmill. Second, we did not include complex 3-D motions at the knee and ankle joints, as we believe that such inclusions would not change the conclusions regarding the joint kinematics and kinetics. Third, we designed the running protocol to assess the acute impact of load carriage on the running biomechanics assuming no impact of muscle fatigue. Therefore, the conclusions may not be valid for prolonged running, which may involve intense muscle fatigue. Fourth, while we performed a power analysis before the start of the study, it is possible that had we used a larger sample size per group, we could have achieved statistical significance of the stature effect on the SF risk. Finally, similar to prior studies [7], [25], we assumed a uniform bone adaptation and repair process when predicting SF risk to the tibia. Future studies may improve the risk prediction model by incorporating muscle fatigue and individualizing the bone adaptation and remodeling mechanisms.

V. Conclusion

Here, we collected experimental data and evaluated the impact of stature and load carriage on the running biomechanics of 21 young, healthy men using computational methods. We found that load carriage, but not stature, significantly changed their running biomechanics. Moreover, we found that, for a 10-week BCT regimen, SF risk increased significantly with heavy load carriage, but not with stature. We expect that the quantitative analysis reported here may help guide training regimens and reduce the risk of SF. In particular, as the Army's new training and testing doctrine shifts from a "one-size-fits-all" to a personalized approach [28], we believe that the ability to perform individualized analysis such as the one described herein will support this doctrine and help enhance peak performance in U.S. Soldiers.

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