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Original article

Influences of load carriage and physical activity history on tibia bone strain

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Abstract

Background: Military recruits are often afflicted with stress fractures. The military's strenuous training programs involving load carriage may contribute to the high incidence of tibia stress fractures in the army. The purpose of this study was to assess the influences of incremented load carriage and history of physical activity on tibia bone strain and strain rate during walking.
Methods: Twenty recreational basketball players and 20 recreational runners performed 4 walking tasks while carrying 0 kg, 15 kg, 25 kg, and 35 kg loads, respectively. Tibia bone strain and strain rate were obtained through subject-specific multibody dynamic simulations and finite element analyses. Mixed model repeated-measures analyses of variance were conducted.
Results: The mean ± SE of the runners' bone strain (μs) during load carriages (0 kg, 15 kg, 25 kg, and 35 kg) were 658.11 ± 1.61, 804.41 ± 1.96, 924.49 ± 2.23, and 1011.15 ± 2.71, respectively, in compression and 458.33 ± 1.45, 562.11 ± 1.81, 669.82 ± 2.05, and 733.40 ± 2.52, respectively, in tension. For the basketball players, the incremented load carriages resulted in compressive strain of 634.30 ± 1.56, 746.87 ± 1.90, 842.18 ± 2.16, and 958.24 ± 2.63, respectively, and tensile strain of 440.04 ± 1.41, 518.86 ± 1.75, 597.63 ± 1.99, and 700.15 ± 2.47, respectively. A dose−response relationship exists between incremented load carriage and bone strain and strain rate. A history of regular basketball activity could result in reduced bone strain and reduced strain rate.
Conclusion: Load carriage is a risk factor for tibia stress fracture during basic training. Preventative exercise programs, such as basketball, that involved multidirectional mechanical loading to the tibia bones can be implemented for military recruits before basic training commences.

Keywords: Modeling; Multi-body dynamics; Strain; Strain rate; Stress fracture; Walking

1. Introduction

Stress fracture is a severe musculoskeletal injury requiring extended periods of recovery time and resulting in significant medical costs.[1−3] Stress fractures are common in sports; the annual injury rate for track athletes has been reported to be 20%.[4] The rate of stress fractures in the military is also high, reaching 6% in the U.S. Army and as high as 31% in the Israeli Army.[1,5] The most common site of stress fractures is in the tibia.[6−8] Tibia stress fractures account for 41%−55% of all stress fractures experienced by athletes involved in running.[9] In the military, tibia stress fractures account for 50% of all stress fractures in male recruits and 35% in female recruits.[6−8] The injury mechanism of stress fractures is not well understood. One hypothesis posits that increased mechanical usage, signified by high strain and high strain rate, stimulates bone remodeling, which results in locally increased porosity and decreased bone mass. Bone mechanical strength is weakened during this stage. When the mechanical loading continues, local stresses are elevated in the bone regions with porosity. If the mechanical loading accumulates at a faster rate than the bone's remodeling process, bone microdamage occurs and microscopic cracks form in the bone,[1] resulting in a stress fracture.[10,11] Thus, stress fractures may be a result of excessively repetitive loads acting on the bone, leading to fatigue-induced bone microdamage.[10,11]

A high volume and intensity of training can result in an increased rate of overuse injuries. During physical training in the military and in sports, a high volume and intensity of training is required for trainees, which can lead to an increase in stress fractures. Stress fracture sites may vary among military trainees and those training for sports. However, tibia stress fracture has been identified as the most common type among

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military trainees and track and field athletes.[6–9] The high incidence of tibia stress fracture in runners is related to the continuous, repetitive muscular activity and ground impacts.[4] Similarly, the high rate of tibia stress fracture in the military is associated with the high volume and high intensity of physical training.[11–13] Loaded long-distance walking and running are major parts of the strenuous training protocol during basic training. The combined mileage of load carriage and running during 12 weeks of basic training can exceed 200 miles.[11]

Several studies have examined the effect of load carriage on walking mechanics.[14–21] Load carriage results in alterations in gait kinematics and kinetics. Kinematically, loaded walking reduces stride length, increases cadence, and increases pelvic tilt and knee and hip flexion.[15,17] Kinetically, load carriage increases ankle, knee, and hip joint moment and power.[21] Studies have also found that walking with additional load results in increased vertical ground reaction forces[14,16,18] in proportion to the amount of load carried.[17–19] In addition, load carriage results in a pronounced increase in ground reaction force loading rate.[20] Increases in ground reaction forces and loading rate combined with increased leg muscle forces present high mechanical stress to the tibia bone. Thus, in theory, tibia bones would be expected to experience elevated strain and increased strain rate during load carriage. However, the effect of load carriage on tibia bone deformation has yet to be determined experimentally. Because incremented load carriage results in increases in ground reaction forces, it is important to examine the tibia bone's response during load carriages by measuring bone strain and strain rate.

Given the significant impact of stress fracture on military and athletic populations, it is essential to determine possible preventative programs. Understanding the effect of past physical activity (PA) on the risk of stress fracture is important. However, findings from the literature are inconclusive. Mustajoki et al.[22] did not find a correlation between past PA and stress fracture risk in the military population. Similarly, Giladi et al.[23] and Swissa et al.[24] could not establish a relationship between army recruits' PA history and the incidence of stress fractures in basic training. However, Gilbert and Johnson[25] reported that recruits with a sedentary lifestyle prior to basic training had a higher risk for stress fracture than those who played sports. Greaney et al.[26] found that Marine recruits who participated in running had fewer stress fracture indications than did nonrunners.[6] Furthermore, Milgrom et al.[27] reported that recruits who played ball sports prior to enlistment had fewer incidences of stress fractures than recruits who did not play ball sports. Because it is evident that engaging in sports involving multidirectional impact loading (e.g., volleyball and gymnastics) results in higher bone mineral density than swimming and running,[28,29] it is speculated that participating in ball sports leads to a general increase of bone mass and strength due to multidirectional impact loadings, which would make tibia bones more resilient to the mechanical loading introduced during basic training. Thus, the risk of stress fracture could be lowered. However, there is no direct evidence that participating in ball sports strengthens tibia bones for load carriage in basic training. It is essential to determine whether ball sports, and the concomitant multidirectional loadings, help reduce bone deformation (strain and strain rate) during load carriage and lessen the chance of developing stress fracture compared with other sports lacking the pattern of multidirectional impacts (e.g., running).

The traditional approach to determining bone strain *in vivo* is to surgically implant strain gauges on bone surfaces. This approach is invasive and results in pain and discomfort. Thus, the number of subjects willing to participate in research examining bone strain has been very small, ranging from 1 to 6 subjects per study.[30–32] In recent years, a noninvasive approach has been implemented to examine bone strain *in vivo*.[33–35] This method incorporates modeling and computer simulation techniques to perform finite element analysis on bone deformation during PAs. Specifically, a human body model is created based on an individual's gender, height, and weight. Bone structures to be examined are modeled based on medical imaging (e.g., computed tomography (CT) and magnetic resonance imaging). These bone models are then transformed into flexible bodies and incorporated into the human body model for multibody dynamic simulations. This flexible multibody dynamic simulation approach has been validated and was found to be comparable to the traditional strain gauge method.[33–35] Given its noninvasive nature, the flexible multibody simulation approach can be used to examine bone strain for large cohorts taking part in PAs.

The purpose of this study was to investigate the influences of incremented load carriage and PA history on tibia bone strain and strain rate during walking by using flexible multibody dynamic simulations. We hypothesized that load carriage would lead to increases in tibia bone strain and in strain rate and that there would be a dose−response relationship between incremented load carriage and bone strain and strain rate. We also expected that PA history would influence the tibia bone strain and strain rate during load carriage. Specifically, we hypothesized that individuals with a history of regular basketball training would have less tibia bone strain and a lower strain rate than those who participated in running.

2. Methods

2.1. Participants

Forty healthy college male participants were recruited to the study. Participants were divided into 2 experimental groups (runners and basketball players). The runner group consisted of 20 recreational runners with a minimum of 2 years of regular running experience (≥15 miles per week). The basketball group consisted of 20 recreational basketball players with a minimum of 2 years of regular basketball playing experience (≥3 times per week). Participants' demographics are presented in Table 1. Participants were recreationally active, classified as low risk for cardiovascular diseases according to American College of Sports Medicine guidelines,[36] and free from known musculoskeletal injury. All participants met the military enlistment standards in terms of physical condition.[37] In addition, the age, body mass, body height, and fitness level maximal oxygen consumption (VO_{2max}) of our participants were

Table 1
Subject demographic information (mean ± SD).

	Runners	Basketball players
Age (year)	20.7 ± 2.4	20.6 ± 1.7
Body height (cm)	180.1 ± 6.3	180.8 ± 8.3
Body mass (kg)	75.7 ± 9.6	80.1 ± 11.5
Leg press 1RM (lb)	492 ± 62	550 ± 127
VO_{2max} (mL/kg/min)	58 ± 7*	50 ± 5

* $p < 0.05$, compared with basketball players.
Abbreviations: 1RM = one-repetition maximum; VO_{2max} = maximum oxygen consumption.

comparable to those of military recruits entering U.S. Army Basic Training.[38] Approvals from the local Institutional Research Board and Human Research Protection Office at the U.S. Army Medical Research and Materiel Command were obtained prior to commencing the study, and all participants signed the informed consent document prior to testing.

2.2. Experimental protocol

The participants completed 3 data collection sessions: (1) assessment of aerobic and muscular strength, (2) motion capture while performing incremented load carriage, and (3) CT imaging of their tibias.

During the aerobic assessment, participants' VO_{2max} was assessed by means of a ParvoMedics TrueOne 2400 metabolic cart (Parvo Medics Inc., Sandy, UT, USA). A modified ramped version of the Bruce treadmill protocol was used.[39] During the muscular strength assessment, the participants' leg muscle strength was measured using a one-repetition maximum (1RM)-effort leg press similar to that described by Hoffman et al.[40]

During motion capture of load carriage, a tandem force-instrumented treadmill (AMTI Advanced Mechanical Technology Inc., Watertown, MA, USA) with 2 force platforms installed under the tandem belts was used to control the walking speed at 1.67 m/s while allowing the ground reaction forces to be detected. Reflective markers were attached on both sides of the body in the following locations to track the walking movement: acromion, sternum, anterior superior iliac spine, posterior superior iliac spine, lateral knee, lateral ankle, heel, base of the fifth metatarsal, and base of the second toe. In addition, 2 cluster marker sets were attached on the thigh and shank, respectively. Fifteen Vicon MX and F-20s series cameras (Vicon Motion Systems Ltd., Denver, CO, USA) were used to track the reflective markers in the space. Vicon Nexus Version 1.4.116 (Vicon Motion Systems Ltd.) was used to collect kinematic data at 240 Hz and ground reaction forces at 2400 Hz.

Participants wore compression shorts, a compression shirt, and cross-training shoes (Proto Speed Lite Trainer; Under Armour, Baltimore, MD, USA) during the experiment. They walked at a self-selected pace on the force-instrumented treadmill for 5 min to warm up. The general experimental protocol consisted of the following walking tasks at a speed of 1.67 m/s: (1) 5 min of normal walking without load carriage, (2) 5 min of walking with a 15-kg rucksack (MOLLE; Specialty Defense System, Dunmore, PA, USA), (3) 5 min of walking with a 25-kg rucksack, and (4) 5 min of walking with a 35-kg rucksack. The loads carried in this study were similar in range to the loads reported in previous studies.[16,41] A 5-min rest break was given between tasks. Ten trials were collected during the 5-min intervals between each walking condition.

During the CT imaging session, axial plane scans of the tibia bones were obtained using a GE Light Speed VCT scanner (GE HealthCare; General Electric Company, Chicago, IL, USA). CT tube potential was set at 120 kv, and the tube current-time product was 144.54 mAs. The slice thickness was set at 0.625 mm. The field of view was 15 cm × 15 cm.

2.3. Computer modeling and simulations

CT images of tibias were segmented in Mimics 14 (Materialise NV, Leuven, Belgium). A surface mesh of the tibia was generated. Solid hexahedral meshing was performed in Marc 2013 (MSC Software Corp., Santa Ana, CA, USA) with an element size of 3 mm^3. Bone density of individual elements was determined based on the average Hounsfield unit value of the pixels contained in the element. Young's modulus was calculated based on the element's density. Poisson ratio was set at 0.3. Individual element's density, Young's modulus, and Poisson ratio were assigned to each element through a custom MATLAB program (Mathworks Inc., Natick, MA, USA). The tibia bone model prepared in Marc 2013 was then converted into a modal-neutral file representing a flexible tibia. This subject-specific flexible tibia model was later used in a forward dynamic simulation of the walking movement to assess tibia bone strain.[33–35]

Motion capture data were processed in Visual 3D Version 4.0 (C-Motion, Inc., Germantown, MD, USA). Kinematic data and ground reaction forces data were filtered using a zero-lag fourth-order Butterworth filter with a cutoff frequency of 8 Hz and 40 Hz, respectively. Subjects' lower extremity musculoskeletal models, including generic geometrics of pelvis, femur, tibia, and foot, were created in LifeMOD 2012 (LifeModeler Inc., San Clemente, CA, USA), which is a plug-in program in ADAMS 2012 (MSC Software Corp.). The generic lower extremity model was then scaled based on the subject's mass, height, gender, and age, as well as the relative positions of the ankle, knee, and hip joints determined from the kinematic data. The flexible tibia represented by the modal-neutral file was introduced to replace the generic tibia in the musculoskeletal model in ADAMS 2012 (MSC Software Corp.) through the LifeMOD plug-in. Tri-axis hinges combined with passive torsional spring-dampers (1 Nm/° and 0.1 Nm·s/°) were employed to model the hip joints. A hinge joint with a single degree of freedom was used for knee and ankle joints in the sagittal plane. A total of 90 muscles were assigned to the legs. The kinematics collected during walking tasks were used to drive the musculoskeletal model with an inverse dynamics algorithm while the muscles' shortening/lengthening patterns were recorded.[33–35] Subsequently, kinematic constraints were removed, and a forward dynamics simulation was performed with muscles serving as actuators and experimental ground reaction forces applied to replicate the walking motion.

Table 2
Effect sizes (partial η^2) associated with strain and strain rate.

Variable	Load carriage effect	Past physical activity effect
Compressive strain	0.553	0.008
Tensile strain	0.467	0.005
Shear strain	0.589	0.011
Compressive strain rate	0.331	0.003
Tensile strain rate	0.312	0.001
Shear strain rate	0.368	0.004

A proportional-integral-derivative feedback controller was implemented to calculate each muscle force magnitude using the error signal between the current muscle length in the forward dynamics and the recorded muscle length during the inverse dynamics simulation.[42–44] The force generated by the individual muscle was limited by its maximum force-generating potential.

Tibia bone strain was then calculated using the Durability plug-in in ADAMS 2012 (MSC Software Corp.). Bone strain and strain rate from the surface of the middle third of the tibia were extracted for analysis. Because other bone strain studies commonly reported the standard error (SE) of the mean to indicate how accurate the estimate of the mean was likely to be,[45–47] SEs were reported along with the means of the bone strain data in this study.

2.4. Statistical analysis

SPSS Version 19.0 (IBM, Armonk, NY, USA) was used to perform statistical analyses. The following dependent variables were examined: peak tensile strain (peak maximal principle strain) and strain rate, peak compressive strain (peak minimal principle strain) and strain rate, and peak shear strain (peak maximal shear strain) and strain rate during the stance of walking. For each of the dependent variables, a mixed model repeated-measures analysis of variance (ANOVA) test was run to determine the effects of subject group (runners *vs.* ball players) and load carriage (4 levels: 0 kg, 15 kg, 25 kg, and 35 kg) on bone strain and strain rate. Statistical effect size (partial η^2) was computed using the following formula: sum of squares of the effect/(sum of squares of the effect + sum of squares of the error).[48] Significance level was set at 0.05.

3. Results

For all the dependent variables examined, significant interactions were found between the subject group and the incremented load carriage ($p = 0.0001$). Therefore, separate ANOVA tests were run to determine the simple effects from the PA history and incremented load carriage on tibia strain and strain rate. Effect sizes associated with each dependent variable are presented in Table 2.

3.1. Tibia bone strain

One-way repeated-measures ANOVA were run to examine the effects of the incremented load carriage on tibia bone strain for the runner group and the basketball player group, respectively. For both groups, significant differences in peak compressive strain, tensile strain, and shear strain (all $p \leq 0.0001$) were found among loaded walking tasks. Specifically, a dose−response relationship between the incremented load carriage and the tibia strain existed. As the load carriage increased, tibia compressive, tensile, and shear strain increased. Table 3 presents the mean ± SE values of the peak compressive, tensile, and shear strain of the middle third of the tibia during loaded walking.

One-way repeated-measures ANOVA were run to examine the effects of PA history on the tibia bone strain for each of the loaded walking tasks. Significant differences in peak compressive strain, tensile strain, and shear strain (all $p \leq 0.0001$) were found between the 2 groups. The runners consistently exhibited greater peak compressive, tensile, and shear strain than the basketball players.

Table 3
Tibia bone strain and strain rate during different loaded walking tasks with load carriages of 0 kg, 15 kg, 25 kg, and 35 kg (mean ± SE).

Variable/Condition	0 kg	15 kg	25 kg	35 kg
Tibia bone strain (μs)				
Runners				
Compressive strain	658.11 ± 1.61##	804.41 ± 1.96**,##	924.49 ± 2.23**,##	1011.15 ± 2.71**,##
Tensile strain	458.33 ± 1.45##	562.11 ± 1.81**,##	669.82 ± 2.05**,##	733.40 ± 2.52**,##
Shear strain	1003.29 ± 2.02##	1229.74 ± 2.40**,##	1444.68 ± 2.72**,##	1586.67 ± 3.48**,##
Basketball players				
Compressive strain	634.30 ± 1.56	746.87 ± 1.90**	842.18 ± 2.16**	958.24 ± 2.63**
Tensile strain	440.04 ± 1.41	518.86 ± 1.75**	597.63 ± 1.99**	700.15 ± 2.47**
Shear strain	972.28 ± 1.96	1155.32 ± 2.32**	1309.96 ± 2.64**	1519.01 ± 3.37**
Tibia bone strain rate (μs/s)				
Runners				
Compressive strain rate	7955.99 ± 24.08#	9477.93 ± 27.04**,##	11,562.32 ± 33.26**,##	11,583.09 ± 37.50
Tensile strain rate	6647.40 ± 20.07##	7901.71 ± 23.27**,##	9551.94 ± 28.28**,##	9617.21 ± 33.14*,##
Shear strain rate	6668.60 ± 13.30##	7828.50 ± 15.27**,##	9594.68 ± 20.05**,##	9662.64 ± 24.26*,##
Basketball players				
Compressive strain rate	7882.92 ± 23.34	9034.16 ± 26.22**	10,293.25 ± 32.23**	11,661.34 ± 36.36**
Tensile strain rate	6557.76 ± 19.46	7432.32 ± 22.57**	8632.39 ± 27.42**	10,012.88 ± 32.13**
Shear strain rate	6540.35 ± 12.90	7488.80 ± 14.80**	8609.44 ± 19.44**	9976.34 ± 23.53**

* $p < 0.05$, ** $p < 0.0001$, compared with previous level of the loaded walking condition; # $p < 0.05$, ## $p < 0.0001$, compared with basketball players.

3.2. Tibia bone strain rate

One-way repeated measures ANOVA were run to examine the effects of the incremented load carriage on tibia bone strain rate for the runner group and the basketball player group, respectively. For both groups, significant differences in compressive strain rate, tensile strain rate, and shear strain rate (all $p \leq 0.0001$) were found between loaded walking tasks. Both the ball players and the runners demonstrated significant increases in peak compressive, tensile, and shear strain rate as load carriage increased from 0 kg to 15 kg, from 15 kg to 25 kg, and from 25 kg to 35 kg. The only exception was that the runners did not show a further increase in compressive strain rate when load carriage was increased beyond 25 kg. Table 3 presents the mean ± SE values of the peak compressive, tensile, and shear strain rate of the middle third of the tibia during loaded walking tasks.

One-way ANOVA tests were run to examine the effect of PA history on tibia strain rate for each of the loaded walking tasks. Significant differences in peak compressive strain rate, tensile strain rate, and shear strain rate (all $p \leq 0.0001$) were found between the runners and basketball players. Specifically, the runners demonstrated greater strain rate than the ball players when carrying 15 kg and 25 kg loads. However, when load carriage was increased to 35 kg, the runners either showed no difference in strain rate (peak compressive strain rate, $p = 0.134$) or showed less strain rate (tensile strain rate and shear strain rate: both $p \leq 0.0001$) compared with the ball players.

4. Discussion

The purpose of the study was to assess the effects of incremented load carriage and PA history on tibia bone strain and strain rate during walking. A group of recreational runners and a group of recreational basketball players participated in the study. Incremented load carriage was introduced at the levels of 15 kg, 25 kg, and 35 kg. Flexible multibody dynamic simulations were performed to obtain tibia bone strain. It was found that significant differences in bone strain and strain rate existed between the runners and the basketball players and also among loaded walking conditions.

We hypothesized that load carriage would increase tibia bone strain. This hypothesis was supported. Compared with the unloaded walking, the 15 kg load carriage resulted in significant increases in tibia bone strain. For runners, the increases were 22%, 23%, and 23% in compressive, tensile, and shear strain, respectively. For ball players, they were 18%, 18%, and 19% in compressive, tensile, and shear strain, respectively. We further hypothesized that a dose−response relationship would exist between incremented load carriage and tibia bone strain. This hypothesis was also supported. We observed that as load carriage was increased to 35 kg from a 15-kg level in 10-kg increments, tibia bone strain continued to rise. High bone strain may trigger a bone remodeling process for building a strong bone to better resist the imposed higher level of mechanical loading. During basic training, military recruits regularly engage in prolonged walking with load carriage. The amount of load carriage has been reported to be as high as 61 kg.[13] The combined loaded walking and running distance during 12 weeks of basic training can reach 200 miles.[11] Repetitive high mechanical loading imposed on the tibia may increase the risk of developing stress fractures. The dose−response relationship between incremented load carriage and bone strain indicates that the risk of stress fracture could be even higher when heavy loads are carried.

We hypothesized that load carriage could lead to an increase in bone strain rate. Our hypothesis was supported. When compared with unloaded walking, carrying a 15-kg load resulted in significant increases in strain rate. For the runners, the increases were 19%, 19%, and 17% in compressive, tensile, and shear strain rate, respectively. For ball players, the increases were 15%, 13%, and 15% in compressive, tensile, and shear strain rate, respectively. We further hypothesized that there would be a dose−response relationship between the incremented load carriage and tibia bone strain rate. This hypothesis was partially supported. We observed that ball players exhibited increases in bone strain rate when load was increased from 15 kg to 35 kg in 10-kg increments. The runners also showed increases in bone strain rate when load was increased from 15 kg to 25 kg. Runners' tensile and shear strain rate continue to increase when load carriage reached 35 kg. However, runners' compressive strain rate remained unchanged when load was increased from 25 kg to 35 kg. Strain rate is a primary mechanical stimulus for bone remodeling.[4,49] Bone is a viscoelastic tissue and is sensitive to strain rate.[50−52] High strain rate could elicit bone remodeling processes for improving bone strength.[2] On the one hand, bone can remodel itself and become stronger to sustain the amount of mechanical load imposed if there is enough time for the remodeling process to be completed. On the other hand, if the bone continues to experience high strain rate in the middle of the remodeling process, microdamage may occur and lead to stress fracture. In this study, the increased strain rate owing to increased load carriage reflected an increased risk of tibia stress fracture. During military basic training, the loaded walking exercise is intense and introduces high strain rate to the tibia bones; thus, it is likely that the risk of tibia stress fracture will be elevated. Interestingly, runners in this study exhibited similar compressive strain rate when carrying 25-kg and 35-kg loads. It is unclear why this might occur. A possible explanation is that when the strain rate reaches a critical level that may be harmful to the bone, the body may regulate forces acting on the tibia to prevent a further increase in the strain rate at specific regions to avoid injury. Based on the current findings, a load carriage of 25 kg or greater may be unsafe to the tibias of runners and may result in a regulation of compressive strain rate. Because the basketball players in this study continued to experience increases in strain rate through the incremented load carriage, it is possible that their tibias were capable of sustaining heavy mechanical loads during load carriage training. Additional studies are needed to examine how strain rate is regulated as load carriage increases.

The effect of PA history on bone quality has been studied on a limited basis. One study reported that athletes engaged in impact-loading sports such as volleyball and gymnastics had higher bone mineral density than swimmers and control subjects.[28] In addition, another study found that runners, swimmers, and divers had deficits in bone mineral density compared with gymnasts and softball players.[29] Furthermore, research demonstrated that a 14-week recreational soccer training program led to greater increases in tibia bone density among participants compared with runners and controls.[53] However, the effect of PA history on tibia bone strain and strain rate has not been fully determined. In this study, we hypothesized that during incremented load carriage the recreational basketball players would show less tibia bone strain compared with the runners. This hypothesis was supported. Compared with the runners, when carrying a 15-kg load, the basketball players exhibited less compressive strain (7% less), tensile strain (8% less), and shear strain (6% less). When carrying a 25-kg load, the ball players showed smaller bone strain in compression (9% less), tension (11% less), and shear (9% less). Furthermore, when load carriage was increased to 35 kg, the ball players continued to show less bone strain in compression (5% less), tension (5% less), and shear (4% less). These results indicate that the history of PA influences tibia bone strain. Individuals with a history of basketball training had less tibia bone strain compared with those with a history of running.

We further hypothesized that the basketball players would show a lower bone strain rate than that of runners during incremented load carriage. This hypothesis was partially supported. Compared with the runners, when carrying a 15-kg load, the ball players showed lower strain rate in compression (5% less), tension (6% less), and shear (4% less). When the load carriage increased to 25 kg, the ball players continued to show less strain rate in compression (11% less), tension (10% less), and shear (10% less). However, when the load carriage reached 35 kg, the runners showed no difference in compressive strain rate and lower strain rate in tension (4% less) and in shear (3% less) than those of the basketball players. It is speculated that a threshold in load carriage exists for individuals. When the load carriage results in a high level of strain rate that might be harmful to the tibia bone, the human body may adjust the individual's walking movement and regulate forces applied to the tibia to lower the strain rate to avoid injury. In this study, the runners may have experienced a threshold of strain rate when carrying a 25-kg load. Because ball players continued to show increases in strain rate when load carriage increased, it appears that the ball players' tibias were resilient to increased strain rate and could sustain mechanical loads of 35 kg or more. Future studies should examine whether the threshold effect of bone strain rate is accompanied with alterations of tibia bone strain distribution.

In this study, we confirmed that past PA has an effect on bone strain and strain rate. Our findings are consistent with previous studies investigating the effect of exercise on bone quality. Running movement introduces cyclic and uniaxial mechanical loading to the tibia bones. Over time, tibia bones adapt to the mechanical loading imposed on them by making some regions of the bone more dense and resilient to loading. However, the overall tibia bone density may not increase much because it is less metabolically costly to maintain a lighter bone. It was reported that a 14-week running exercise program did not effectively increase tibia bone density.[53] However, a 14-week soccer training program did result in a significant increase in tibia bone density.[53] It was surmised that the movements used in soccer, such as acceleration, deceleration, jumping, and cutting, produced mechanical stimuli to tibia bones from different directions and contributed to changes in bone mass.[53] Similar to soccer, basketball introduces multiaxial mechanical loading to the tibia bone and could lead to positive osteogenesis along the whole bone. Subsequently, overall bone strength is improved. In this study, we observed that basketball players' tibia bones were stronger and that the players experienced less strain and a lower strain rate in their tibias than did runners during load carriage.

In this study, we recruited recreational runners and basketball players. In general, these participants did not participate in routine weight-training exercises. Although there was no statistical difference in maximal leg press between the groups, the greater leg strength exhibited by the ball players could be a result of playing basketball regularly, which comprises running, jump-landing, cutting, and sudden accelerations and decelerations of the body. In addition, the greater VO_{2max} associated with runners appeared to be a result of their habitual running activity.

5. Conclusion

In summary, load carriage resulted in increases in bone strain and strain rate. A dose−response relationship existed between incremented load carriage and bone strain and strain rate. History of PA influenced bone strain and strain rate during load carriage. A history of regular basketball activity resulted in reduced bone strain and strain rate. Therefore, military administrators and recruits need to be aware that load carriage is a risk factor for tibia stress fracture during basic training. Preventative exercise programs involving multidirectional mechanical loading to the tibia bones, such as basketball and soccer, should be implemented for recruits before basic training. Furthermore, track coaches and distance runners should recognize the importance of strengthening the tibia bones to reduce the incidence of stress fractures. Runners should consider regularly engaging in exercises that involve multiaxial mechanical loadings to the lower extremity bones.

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Authors' contributions

HW carried out the study, participated in data collection, data processing, and data analysis, and drafted the manuscript; MK participated in data processing and data analysis and helped draft the manuscript; DCD participated in data collection and data analysis and helped draft the manuscript. All authors have read and approved the final version of the manuscript, and agree with the order of presentation of the authors.

Competing interests

The authors declare that they have no competing interests.

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