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Original research

The relationships between multiaxial loading history and tibial strains during load carriage

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ABSTRACT

Objectives: To determine if a history of exercise involving multiaxial loading, through soccer participation, influences tibial stains during incremented load carriage.
Design: Cross-sectional study.
Methods: 20 female soccer players (20 ± 1 yr) and 20 mass- and height-matched healthy women (21 ± 1 yr) participated in walking tasks with 0 kg, 10 kg, 20 kg, and 30 kg loads on a force instrumented treadmill at 1.67 m/s. Subject-specific tibial CT models were combined with subject-specific musculoskeletal models for forward-dynamic simulations and finite element analyses. Strains from the middle third of the tibial shaft were analyzed. A mixed model repeated measures analysis of variance (ANOVA) and one-way ANOVAs were run with a Bonferroni correction setting significance at 0.0009.
Results: Significant differences in tibial characteristics were found among loading conditions and between groups (all p < 0.0001). Tensile strains were 19.6%, 22.2%, 44.1%, and 20.7% lower in soccer players at 0 kg, 10 kg, 20 kg, and 30 kg of load carriage, respectively. Strain rates were 20.4%, 29.9%, 43.4%, and 18.9% lower, respectively, in soccer players. Lower compressive and shear strain magnitudes and rates were also observed in soccer players, with the only exception at the 30 kg loading condition in which controls had 2.4% lower strain magnitudes in compression, on average, compared to soccer players.
Conclusions: A history of activity involving multiaxial loading was associated with generally lower estimated tibial strains during load carriage. Lower strain levels during repetitive physical activity may be protective from stress fracture. These findings suggest that physical training, such as participating in soccer, may be effective for preconditioning prior to entering military or endurance training.

1. Introduction

Stress fractures are injuries that occur with repetitive mechanical loading of the skeleton and are therefore common in military trainees and endurance athletes. With up to a 20% incidence of stress fracture in female Soldiers undergoing military training and with a seasonal incidence of up to 20% in female endurance runners, these physically active female populations suffer stress fractures at rates 3–4 times higher than those of their male counterparts.[1,2] In the military, stress fractures occur most frequently in the lower extremities, with the most common fracture sites at the tibia, followed by the fibula, metatarsals, femur, and pelvis.[3] In athletes, the most common sites are the metatarsals, tibia, lower back, lumbar spine, and pelvis regions.[4]

Stress fractures during military training have been principally attributed to repetitive loading from tasks such as marching with load carriage, running, and other physical military drills.[1] When bones are repetitively loaded, the strain or deformation of the bone tissue has been shown to contribute to new bone deposition on existing surfaces. In turn, this new bone formation can favorably influence skeletal fatigue resistance.[5] However, the strains that stimulate new bone deposition may also generate fatigue damage within the bone tissue in the forms of diffuse matrix damage and microcracking.[6] This latter mode of damage, in the absence of adequate time for tissue self-repair, has been posited to coalesce upon continued loading—ultimately leading to stress fracture.[6] Accordingly, introducing pretraining exercise programs, aimed at stimulating bone formation and therefore reducing bone strains during endurance and military activities, has been suggested for stress fracture prevention.[7]

Of potential modalities for prophylactic exercise, sports such as soccer, volleyball, and basketball may be particularly effective.

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These sports include cutting and multidirectional movements that likely generate multiaxial loading of the tibia, which could theoretically stimulate new bone deposition to a greater degree than more uniformed patterns of loading such as running. In alignment with this theory, ball sports have been associated with favorable bone mechanical properties.[8] For example, elite soccer players were reported to have higher whole body and regional bone mineral density (BMD) at the spine, hip, and calcaneus compared to sedentary controls and higher hip and spine BMD than elite runners.[8] These favorable bone characteristics could conceivably result in lower strains during loading. The magnitude and rate of strain with loading are important determinants of bone fatigue and thus, may influence stress fracture risk.[9] This notion is supported by observations that a history of playing ball sports is associated with reduced risk of stress fracture in Soldiers and athletes.[10,11] However, whether strains during loading are lower in those with a history of ball sports is unknown. Measuring *in vivo* bone tissue strains has classically been conducted with surgical strain gauge placement on the exterior surface of a bone[12] which is limited by the invasive nature of the procedure. Nonetheless, biomechanical models employing bone imaging, motion capture, and finite element analysis have been shown to provide non-invasive estimates of tibial strain characteristics.[13–15]

Therefore, the purpose of this study was to use a biomechanical model to determine if tibial strain characteristics during walking with load carriage differ between women with a history of soccer and sedentary controls. We chose to perform the study in women, given higher rates of stress fracture compared to their male counterparts, and we hypothesized that women with a history of soccer would experience lower tibial strains and strain rates for a relative loading condition compared to sedentary controls.

2. Methods

Forty healthy college female participants were recruited. Twenty intercollegiate soccer players with a minimum of two years of regular soccer training formed the soccer group, and 20 mass- and height-matched healthy women formed a control group. Participants were classified as low risk for cardiovascular diseases according to American College of Sports Medicine (ACSM) guidelines[16] and free from known musculoskeletal injury. Participants in the control group were identified as sedentary in the two years prior to the study, based on ACSM guidelines (<150 min of moderate exercise per week).[16] All participants met military enlistment standards in terms of physical condition, and age, body mass, and body height of participants were comparable to those of military recruits entering US Army Basic Training.[17] Approvals from the local Institutional Research Board were obtained prior to commencing the study.

Participants completed two data collection sessions: (1) motion capture while performing incremented load carriage, and (2) tibial CT imaging. During motion capture, a tandem force instrumented treadmill (Advanced Mechanical Technology, Inc., Watertown, MA, USA) with 2 force platforms installed under the tandem belts was used to control walking speed at 1.67 m s^{-1} while allowing ground reaction forces to be detected. Reflective markers were attached to the sternum and to both sides of the body in the following locations to track walking movement: acromion, 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, 4-marker cluster sets were attached on the thigh and shank, respectively. Fifteen Vicon^{TM} MX and F-20s series cameras (Vicon Motion Systems Ltd., Denver, CO, USA) were used to track the reflective markers in space. Vicon NEXUS^{TM} (V 2.4) (Vicon Motion Systems Ltd., Denver, CO, USA) was used to collect kinematic data at 200 Hz and ground reaction forces at 2000 Hz.

Participants wore compression shorts and shirt and military combat boots (Rocky S2V, USA) and walked at a self-selected pace on the treadmill for 5 min to warm up. The experimental protocol consisted of the following walking tasks at a speed of 1.67 m/s for 2 min: (a) normal walking without load carriage, (b) walking with a 10 kg rucksack (MOLLE; Specialty Defense System, Dunmore, PA, USA), (c) a 20 kg rucksack, and (d) a 30 kg rucksack. The loads carried were similar in range to those reported in a prior study.[18] A 5-min rest was scheduled between tasks. Walking tasks were randomized. Three, 7-s trials were collected during the 2-min interval of each walking condition, with 6–7 gait cycles included in each 7-s trial. A total of five gait cycles, defined as heel strike of one foot to the next heel strike of the same foot, were analyzed per participant.

CT imaging consisted of axial plane scans of the tibia using a GE Light Speed VCT scanner (GE HealthCare, General Electric Company, Chicago, IL, USA). CT tube potential was set at 120 peak kilovoltage (kVp), and the tube current-time product was set at 144.54 milliampere-seconds (mAs). The slice thickness was 0.625 mm. The field of view was 15 cm × 15 cm.

Tibial CT images were imported and segmented in Mimics 17 (Materialise NV, Leuven, Belgium). A surface mesh of the tibia was generated. Solid hexahedral meshing was performed in Marc 2016 (MSC Software Corporation, 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.[19] Individual element density, Young's modulus, and Poisson ratio were assigned to each element through a custom MatLab program (Mathworks Inc., Natick, MA, USA). The tibial model was then converted into a modal-neutral file (MNF) representing a flexible tibia. This subject-specific flexible tibial model was later used in a forward dynamic simulation of the walking movement in order to assess tibial strain (Fig. 1).[15] For discussion purposes, we also evaluated the stride length and cadence during the various loading conditions.

Motion capture data were processed in Visual 3D^{TM} v5.0 (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. Subject-specific lower extremity musculoskeletal models including generic geometrics of pelvis, femur, tibia, and foot were created in LifeMOD 2012 (LifeModeler Inc., San Clemente, CA, USA), a plug-in program in ADAMS 2012 (MSC Software). The generic lower extremity model was then scaled based on the participant's mass, height, sex, 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 MNF file was introduced to replace the generic tibia in the musculoskeletal model in ADAMS 2012 (MSC Software Corporation, Santa Ana, CA, USA) through the LifeMOD Plug-in. Spherical, revolute, and universal joints were used to model the hip, knee, and ankle joints, respectively. Passive torsional spring-dampers (1 Nmm/° and 0.1 Nmm s/°) were assigned to the joints. 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 kinematics algorithm while the muscles' shortening/lengthening patterns were recorded.[15] 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.

A proportional-integral-derivative feedback controller was implemented to calculate each muscles force magnitude using the error signal between the current muscle length in the forward dynamics and the recorded muscle length during the inverse

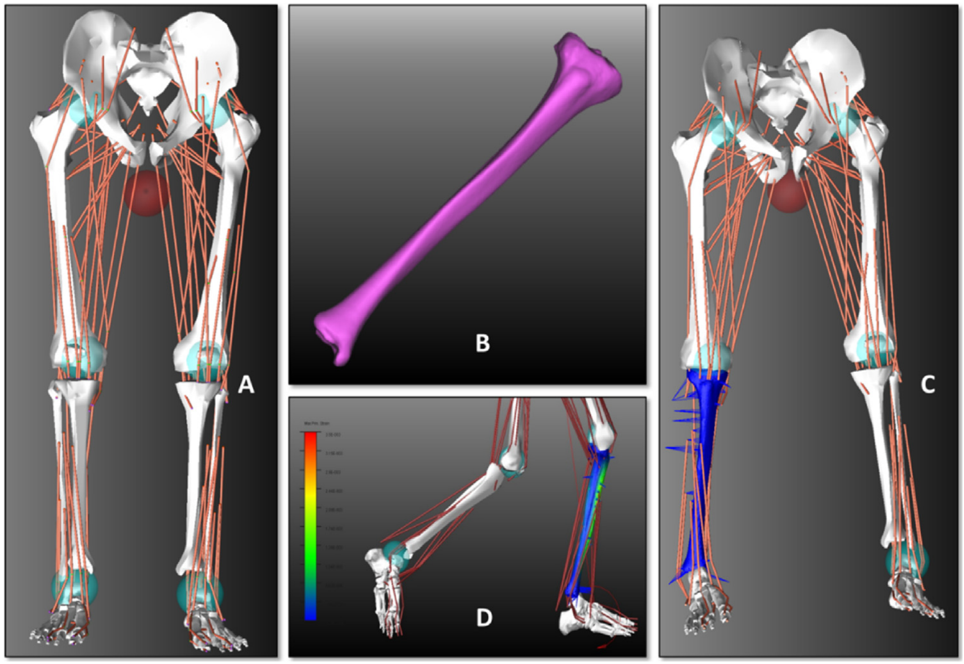

Fig. 1. (A) Lower-body musculoskeletal model consisting of 7 segments and 90 muscles. (B) Subject-specific tibia obtained from CT scan. (C) A subject-specific musculoskeletal model including subject-specific tibia. (D) An instance of a forward dynamic multi-body computer simulation.

dynamics simulation.[20,21] 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. Bone strain magnitudes and strain rates in tension, compression, and shear were extracted from the surface of the entire middle third of the tibial shaft for analysis.

SPSS v 24 (SPSS Inc. Chicago, IL, USA) was used to perform statistical analyses. The following dependent variables were examined: peak tensile strain (peak maximal principal strain) and strain rate, peak compressive strain (peak minimal principal strain) and strain rate, and peak shear strain (peak maximal shear strain) and strain rate during the stance of walking. A mixed model repeated measures analysis of variance (ANOVA) was run to evaluate the significance of the interaction between a history of soccer playing and incremental increase in load on strain. When the interaction was significant, we then conducted a series of one-way ANOVAs to examine the effects of increasing amounts of load carriage (0 kg vs. 10 kg, 10 kg vs. 20 kg, and 20 kg vs. 30 kg) on tibial strain characteristics for the soccer and control groups, respectively. We then conducted one-way ANOVAs to examine the effects of multi-axial loading history (soccer vs. control group) on the tibial strain magnitudes and rates for each of the load carriage walking tasks (at each of the 4 levels: 0 kg, 10 kg, 20 kg, and 30 kg). To account for the multiple ANOVAs performed, a Bonferroni correction was performed, and significance level was set at 0.0009.

3. Results

Forty participants completed the study. As designed, the 20 participants in the soccer group were, on average, of similar age (20 ± 1 yr vs. 21 ± 1 yr), height (166 ± 5 cm vs. 166 ± 4 cm), and mass (62 ± 6 kg vs. 61 ± 10 kg) as the 20 participants in the control group, respectively. In the soccer group, 19 participants self-reported as Caucasian and one as Black. In the control group, 14 participants self-reported as Caucasian and 6 as Asian. All participants completed the four loading conditions (0 kg, 10 kg, 20 kg, and 30 kg).

Significant interactions were found between the subject group and incremented load carriage for all dependent variables examined (all $p < 0.0001$). This is supported by comparing strain characteristics at the various loading conditions between soccer players and controls, where we observed significant differences in strain magnitudes and rates between the two groups (all $p \leq 0.0001$). The soccer group consistently exhibited lower peak compressive, tensile, and shear strain magnitudes and rates compared to controls during all loading conditions except the 30 kg loading condition at which the control group had lower compressive strain magnitudes (Tables 1 and 2 and Fig. 2).

Furthermore when each group was evaluated separately, significant differences were observed in peak compressive, tensile, and shear strain magnitudes and rates from one loading condition to the next (from 0 kg to 10 kg, 10 to 20 kg, and 20 kg to 30 kg, all $p < 0.0001$). Specifically, the peak strain magnitudes and rates significantly increased incrementally with increases in the amount of load carried from 0 kg to 30 kg in the soccer group and from 0 kg to 20 kg in the control group. However, strain magnitudes and rates at the 30 kg load were lower than those of the 20 kg load in the control group.

Both groups exhibited similar trends in changes in stride length and cadence during incremented load carriage. However, the rates of change in stride length and cadence was different between the groups. In particular, when load carriage changed from 20 kg to 30 kg, the soccer group showed a 2.6% decrease in stride length and

Table 1
Peak tibial bone strains during walking with load carriages of 0 kg, 10 kg, 20 kg, and 30 kg (mean ± SE).

Tibia bone strain (μs)	0 kg	10 kg	20 kg	30 kg
Soccer group				
Compressive strain	849 ± 3#	960 ± 4*,#	1092 ± 10*,#	1180 ± 6*,#
Tensile strain	562 ± 2#	634 ± 3*,#	736 ± 7*,#	825 ± 4*,#
Shear strain	651 ± 2#	736 ± 2*,#	838 ± 7*,#	922 ± 4*,#
Control group				
Compressive strain	1039 ± 3	1179 ± 4*	1749 ± 11*	1152 ± 7*
Tensile strain	684 ± 3	792 ± 3*	1152 ± 7*	1015 ± 4*
Shear strain	799 ± 2	908 ± 2*	1325 ± 7*	1200 ± 4*

* $p \leq 0.0001$, compared with previous level of the loaded walking condition.
\# $p \leq 0.0001$, compared with controls.

Table 2
Tibia bone strain rates during walking with load carriages of 0 kg, 10 kg, 20 kg, and 30 kg (mean ± SE).

Tibia bone strain rate (μs/s)	0 kg	10 kg	20 kg	30 kg
Soccer group				
Compressive strain rate	$12,757 \pm 61^{\#}$	$13,549 \pm 66^{*,\#}$	$16,486 \pm 253^{*,\#}$	$17,744 \pm 89^{*,\#}$
Tensile strain rate	$10,509 \pm 55^{\#}$	$11,076 \pm 59^{*,\#}$	$13,147 \pm 119^{*,\#}$	$14,525 \pm 70^{*,\#}$
Shear strain rate	$10,879 \pm 37^{\#}$	$11,514 \pm 41^{*,\#}$	$13,803 \pm 161^{*,\#}$	$15,086 \pm 53^{*,\#}$
Control group				
Compressive strain rate	$16,279 \pm 64$	$18,840 \pm 70^{*}$	$31,433 \pm 267^{*}$	$23,530 \pm 94^{*}$
Tensile strain rate	$12,890 \pm 58$	$14,964 \pm 62^{*}$	$20,425 \pm 125^{*}$	$17,558 \pm 74^{*}$
Shear strain rate	$13,746 \pm 39$	$15,917 \pm 43^{*}$	$24,168 \pm 170^{*}$	$19,345 \pm 56^{*}$

* $p \leq 0.0001$, compared with previous level of the loaded walking condition.
\# $p \leq 0.0001$, compared with controls.

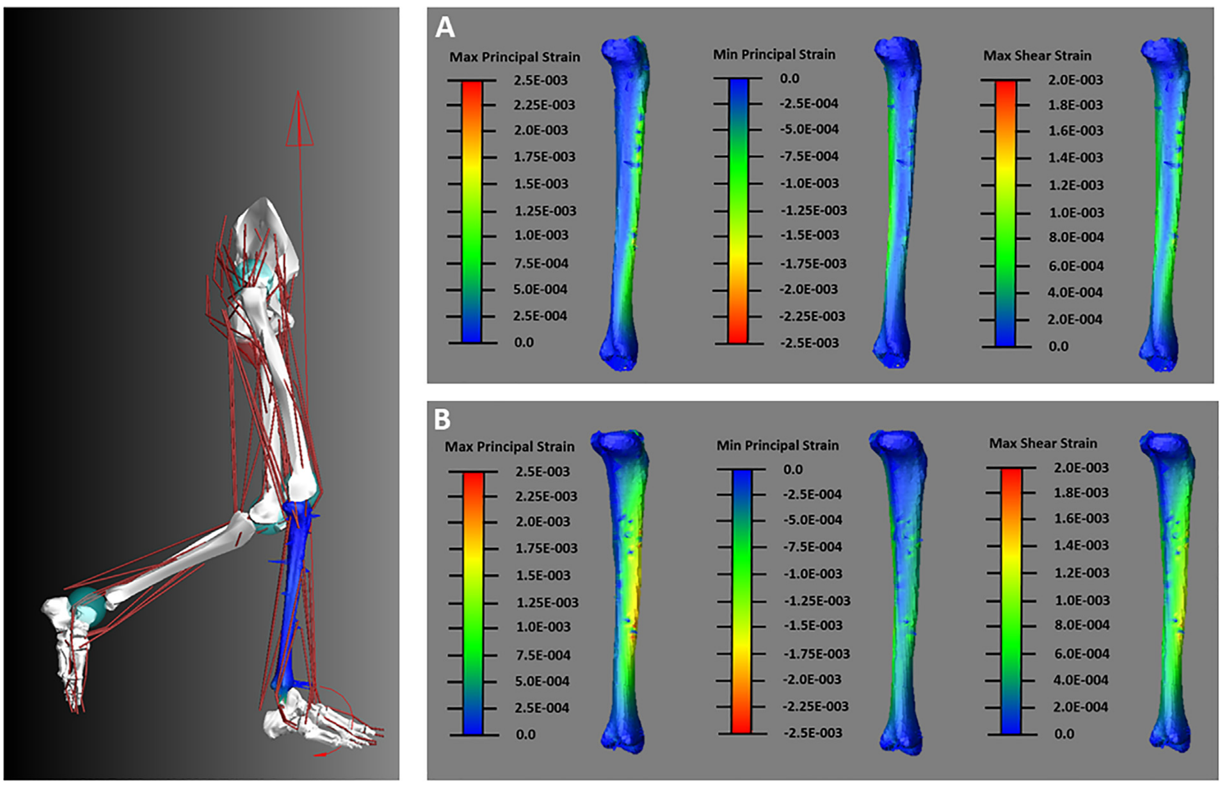

Fig. 2. Sample images of tibial bone strains during mid-stance of load carriage (10 kg). Row A shows a soccer player's bone. Row B shows a healthy control's bone. Maximal principal strain, minimal principal strain, and maximal shear strain are presented from left to right, respectively.

a 2.3% increase in cadence, while the control group showed a 4.6% decrease in stride length and a 4.5% increase in cadence.

4. Discussion

In this study of tibial strain characteristics during walking with incrementally increasing load carriage, we found that a history of soccer participation, a sport that involves multiaxial loading, was associated with 20–46% lower strain magnitudes and 19–62% lower strain rates compared to controls for nearly all loading conditions. As stress fracture from fatigue of bone tissue is a function of strain magnitude and strain rate multiplied by the number of loading cycles,[9] these findings suggest that a history of multiaxial loading may provide mechanical advantages at the tibia.

The only exception to the pattern of lower strain magnitudes in soccer players was a 2.4% lower strain magnitude, in compression only, in the control group compared to the soccer group at the 30 kg loading condition. We also observed an unexpected reduction in strain magnitudes and rates from the 20 kg to the 30 kg loading conditions in controls only. All other loading increments resulted in proportional increases in tibial strain characteristics in both groups. The greater strain magnitude in soccer players compared to controls at 30 kg of load carriage and the reduced strain magnitudes and rates from 20 kg to 30 kg in the control group could potentially reflect a threshold whereby increasing loads to 30 kg or more may stimulate alterations in gait, such as taking shorter steps.[22] Indeed, we did observe a greater decrease in stride length and greater increase in cadence in the control group than in the soccer group, which supports this concept. These findings support the notion that, beyond the characteristics of the loading environment, the management of these loads may be an important factor in injury risk.[23]

Findings of predominantly proportional increases in strains with increasing load carriage are congruent with reports that ground reaction forces also increase proportionately with increasing loads.[24,25] For example, increasing load carriage by 15% of body weight resulted in a comparative 15% increase in ground reaction forces in young adults.[25] While these studies suggest that external forces on the bones of the lower extremities increase in parallel with the magnitude of load carriage, our findings, combined with similar observations of increasing tibial strains with increasing load carriage in a single individual[14] and in recreational athletes,[24] suggest that the increases in external forces upon the tibia generally translate to proportional increases in internal bone strains. These studies all support the concept that greater external loads borne by Soldiers have direct mechanical consequences within the tibia during walking with load carriage. Consequently, efforts to minimize the loads carried by Soldiers or identify means of lessening the burden of these loads on the musculoskeletal system may diminish risk of stress fracture during military training and operations.

While strain characteristics predominantly increase in proportion to the magnitude of load carriage, our findings that these strain characteristics are appreciably lower in those with a history of participation in soccer suggest that activities that involve multiaxial loading may attenuate the bone strains generated during repetitive loading that have been posited to contribute to stress fracture development.[6] These conclusions are supported by a recent report that estimated tibial strains were significantly lower in basketball players compared to endurance runners during load carriage.[15] The lower *in vivo* strains reported in basketball and soccer players provide at least a partial mechanical basis for the 82% and 84% reductions in stress fracture risk observed in Soldiers and runners, respectively, with a history of participation in ball sports.[7,26] These lower strains may be rooted in the favorable bone density and morphology parameters observed in those participating in ball sports.[27] Regardless of the mechanisms, collectively, this growing body of literature provides evidence for the potential benefits of initiating a preconditioning program, involving progressive introduction of multiaxial loading, prior to endurance or military training to offset risk of stress fracture. Nevertheless, until a more uniformed loading condition such as running is investigated under similar experimental conditions, the direct effects of multiaxial loading above those of mechanical loading alone cannot be fully determined.

Strengths of this study include subject-specific CT scans and musculoskeletal models to non-invasively estimate tibial strains in a reasonably large cohort. Tibial strain magnitudes and rates predicted by this model were consistent with values reported in the literature from studies of direct *in vivo* measurements.[7,28] In these reports, tensile strains of the tibia during walking was reported in the range between 394–840 microstrains, and compressive strains of the tibia ranged from 434 to 672 microstrains. The current study showed that tensile strains ranged from 562 to 648 microstrains and compressive strains ranged from 849 to 1039 microstrains during walking without loads. Strain differences between this study and prior studies may be due to differences in sex (female subjects in current study compared to male subjects in other studies) and walking speed (1.67 m/s in current study compared to slower speed ranged from 1.39 m/s to 1.67 m/s in other studies).

While participants were well-matched in body size and age, one potential limitation of this study is the differing distribution of races between groups. Specifically, the soccer group had a greater proportion of Caucasian participants, and the control group a greater proportion of Asian participants. Stress fracture risk varies by race in the U.S. Army population,[29] with Caucasian Soldiers possessing greatest risk, Asian Soldiers at an intermediate risk, and Black Soldiers with the lowest risk of stress fracture. Racial differences in bone mechanical properties appear to underpin some of these differences in risk, with relatively poorer indices of cortical area, vBMD, trabecular thickness, and bone strength reported in Caucasian men and women compared to Black men and women.[30] Therefore, because there were more Caucasian women in the soccer group, we would expect strains to be relatively higher on average if there was no effect of mechanical loading history. To the contrary, we observed attenuated strains in the soccer groups which suggest that a history of mechanical loading may indeed confer mechanical advantages at the tibia. Another limitation of the study is that we cannot control for potential selection bias in which soccer players may select to play their sport because they are generally healthy and have favorable tibial mechanical properties that put them at low risk for injury. More favorable tibial properties would then manifest as lower strains during load carriage. Until prospective studies are conducted demonstrating improvements in indices of bone strength and subsequent decreases in bone strains with multiaxial loading, compared to a more uniformed loading sport as well as a control group, the direct contribution of this loading modality to tibial loading characteristics cannot be discerned. Finally, in this study, we did not account for potential differences between groups in characteristics and behaviors known to influence skeletal health, including menstrual history, physical activity levels during youth, and current and past dietary practices.

5. Conclusion

We found that a history of soccer participation was associated with generally lower estimated tibial strains during load carriage relative to healthy but sedentary controls. Lower strains during repetitive physical activity may protect from stress fracture. These findings suggest that physical training involving multiaxial loading may be an ideal modality for preconditioning prior to entering military or endurance training, although this remains to be determined.

Practical implications

- Participants with a history of soccer have shin bones that are more resistant to deformation during loading.
- Bones that are more resistant to deformation with loading may be at lower risk of stress fracture.
- These results suggest that introducing pretraining programs, prior to endurance sport or military training, may reduce risk of stress fracture.

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