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Dual-Energy X-Ray Absorptiometry Derived Structural Geometry for Stress Fracture Prediction in Male U.S. Marine Corps Recruits*

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ABSTRACT

A total of 626 U.S. male Marine Corps recruits underwent anthropometric measurements and dual-energy X-ray absorptiometry (DXA) scans of the femoral midshaft and the distal third of the tibia prior to a 12 week physical training program. Conventionally obtained frontal plane DXA scan data were used to measure the bone mineral density (BMD) as well as to derive the cross-sectional area, moment of inertia, section modulus, and bone width in the femur, tibia, and fibula. During training, 23 recruits (3.7%) presented with a total of 27 radiologically confirmed stress fractures in various locations in the lower extremity. After excluding 16 cases of shin splints, periostitis, and other stress reactions that did not meet fracture definition criteria, we compared anthropometric and bone structural geometry measurements between fracture cases and the remaining 587 normals. There was no significant difference in age ($p = 0.8$), femur length ($p = 0.2$), pelvic width ($p = 0.08$), and knee width at the femoral condyles ($p = 0.06$), but fracture cases were shorter ($p = 0.01$), lighter ($p = 0.0006$), and smaller in most anthropometric girth dimensions ($p < 0.04$). Fracture case bone cross-sectional areas ($p < 0.001$), moments of inertia ($p < 0.001$), section moduli ($p < 0.001$), and widths ($p < 0.001$) as well as BMD ($p < 0.03$) were significantly smaller in the tibia and femur. After correcting for body weight differences, the tibia cross-sectional area ($p = 0.03$), section modulus ($p = 0.05$), and width ($p = 0.03$) remained significantly smaller in fracture subjects. We conclude that both small body weight and small diaphyseal dimensions relative to body weight are factors predisposing to the development of stress fractures in this population. These results suggest that bone structural geometry measurements derived from DXA data may provide a simple noninvasive methodology for assessing the risk of stress fracture. (J Bone Miner Res 1996;11:645–653)

INTRODUCTION

Stress fractures cause significant morbidity among athletes and during military recruit training, particularly for elite programs requiring intense physical conditioning such as the U.S. Marine Corps.[1–3] Data indicate that the incidence of stress fracture during recruit training occurs in approximately 3–4% of Marine recruits over the initial 12 week training period.[2,4] Previous analyses have shown that the primary fracture site is the tibia, followed by metatarsals, femur, and calcaneus in descending order of prevalence.[2]

Stress fractures, which predominantly occur in the lower extremity, are believed to result from bone structural failure caused by repetitive weight bearing loads. Weight bearing under training regimens subjects these bones to repetitive

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axial compression, torsion, and bending stresses.[5] Within a bone under a given load, stress magnitudes are determined by bone structural geometry, while the bone's ability to resist these stresses is defined by bone material properties.[6,7] Since bone material properties are much less variable than structural geometry, it is likely that most of the individual differences in bone strength are geometric.[7]

For a given long bone, the most important geometric properties are the cross-sectional area (CSA), and for bending and torsion the cross-sectional moments of inertia (CSMI). Structural properties of the long bones of the lower extremity are strongly body size–dependent and are known to vary with age and gender.[8–10] For example, previous work in our laboratory suggests that gender differences in elderly fracture rates may relate to the ability of bone to alter the cross-sectional moments of inertia to compensate for increased mechanical stresses due to bone loss with aging.[11] There is also evidence that bone is structurally remodeled over shorter time scales to restrict stresses in limbs subjected to increased loads; moreover, these changes are evident in the cross-sectional properties.[12] Giladi and colleagues[13] in a prospective study of 295 Israeli infantry recruits showed that those with narrow tibias had higher stress fracture rates than those with wider tibias. Bone width is mathematically related to the cross-sectional moment of inertia and hence can be used as an index of strength in bending and torsion. Later work by the same group[14] confirmed that stress fractures are best determined by measurement of tibial cross-sectional moments of inertia.

A technique for deriving cross-sectional geometric properties of long bones from bone mineral data acquired with absorptiometry scanners has been previously described.[15,16] In this investigation, we apply the method to a prospective study of a large (>600) sample of male U.S. Marine Corps recruits followed through a 12 week training program. Bone mineral density (BMD) and cross-sectional geometric properties of the midshaft femur and mid-distal tibia were determined from dual-energy X-ray absorptiometry (DXA) data for each subject at enrollment. Various anthropometric data were also collected for each recruit. These data are used to explore further the relationship between bone structural properties, body size, and subsequent stress fracture incidence in this sample.

MATERIALS AND METHODS

Subjects

This prospective study followed 626 Marine Corps recruits for 12 weeks of basic training at the U.S. Marine Corps Recruit Depot in San Diego, California, who arrived between January 1993 and April 1994. A random sample totaling approximately 5% of the recruits was selected for study. Following informed consent, DXA scans and anthropometric measurements were obtained.

TABLE 1. ANTHROPOMETRIC MEASUREMENTS TAKEN FOR EACH OF THE 626 MARINE CORPS RECRUITS AT THE TIME OF DXA SCANNING, PRIOR TO 12 WEEKS OF BASIC TRAINING

Measurement	*Definition*
Femur length	Distance from anterior superior iliac spine (ASIS) to tibial tuberosity
Tibia length	Distance from tibial tuberosity to medial malleolus
Bi-iliac breadth	Horizontal distance between pelvic crests
Femur bicondylar breadth	Mediolateral width of distal femur between femoral condyles
Neck girth	Measured at middle of neck
Waist girth	Measured at level of pelvic crests
Thigh girth	Measured at midpoint between ASIS and tibial tuberosity
Calf girth	Measured at widest part of calf

Anthropometric measurements

Height and weight were measured for each recruit at the time of scanning. The body mass index (BMI) was calculated as weight/height^2, with weight in kilograms and height in meters. Several other anthropometric measurements were also taken which are listed with their definitions in Table 1. Bone lengths were used to identify DXA scanning locations and as measurements of limb segment lengths. The bicondylar breadth of the femur is a measurement of joint size. Other breadths and girths provide indices of skeletal frame size and soft tissue distribution. All measurements were taken to the nearest millimeter. Lower limb dimensions were taken on both the right and left sides, but only data for the right side (the side on which DXA scans were obtained) are reported here. Femur and tibia lengths were measured during the scanning procedure as described below. Other dimensions were recorded immediately before scanning using either calipers (breadths) or a tape measure (girths and lengths).

DXA scanning

Scanning was done with a conventional DXA scanner (Norland XR26, Norland Inc., Fort Atkinson, WI, U.S.A.). Subjects were positioned supine on the scanning table. Only the right side was scanned at both the midthigh (midfemur length) and distal third of the lower leg (one-third tibial length from the medial malleolus). The femur length was measured with a tape measure from the anterior-superior iliac spine to the proximal edge of the tibial tuberosity (which provides a reproducible but approximately 5% overestimate of actual femur length). The tibia length was measured as the distance from the proximal edge of the tibial tuberosity to the most medially projecting point of the medial malleolus. Femur scans were done with the plane of knee joint flexion perpendicular to the scan table. Lower leg scans were done with the leg fixed with Velcro straps in a jig which cradles the foot and ankle rotated 17° medially about the leg axis. This position was used to permit a clear distinction between the tibia and fibula when viewed anteroposteriorly.[17] Both scan regions were approximately 0.5 cm long by 10–15 cm wide, oriented across the long axis of the bone, which in turn was aligned by eye along the scan table long axis. Each scan region included approximately 10 scan lines traversing the long axis of the bone. The scan speed was set to 10 mm/s with a pixel spacing of 0.5 × 0.5 mm. The total scan time was 3–4 minutes/location.

Prior to initiation of the study subject accrual, scan precision was measured with five repeated scans at both the thigh and lower leg locations on 10 volunteer subjects. Between scans, subjects were removed from the table and then repositioned. Periodically during the study, DXA scan data were transferred onto diskette files to a separate computer for analysis, using a technique previously applied to the proximal femur[11,16,18] modified for use with diaphyseal sections. While details of the general methodology have been given elsewhere,[15,16] the method is briefly described in the Appendix. A computer program is employed to extract profiles of bone mass traversing the bone from the bone mineral scan data. These profiles of bone mass are used to derive bone cross-sectional areas, bone widths, and cross-sectional moments of inertia in the scanning plane. In the current study, we determined these properties individually for each of the (≈10) scan lines traversing the bone axis and then reported the average value.

The cross-sectional area is an index of axial strength and is related to shear strength, while the cross-sectional moment of inertia is an index of bending rigidity. Subperiosteal bone width and conventional bone mineral "density" (BMD) are also calculated. The section modulus (Z), an index of bending strength, was also calculated as cross-sectional moment of inertia divided by half bone width. For the lower leg scan data, the bone mineral and geometric section properties were determined separately for the tibia and fibula.

Injury tracking

Subjects included in the study were passively followed longitudinally through the 12 week training period, and all injuries self-reported to the Sports Medicine Clinic were recorded in a computer database using standard ICD-9-CM Expanded Orthopaedic classifications.[19] For purposes of this study, stress fractures were defined as a partial or complete fatigue fracture of insidious onset in nondiseased bone. Diagnosis of stress fractures was based on (1) clinical presentation of localized pain of insidious onset, without prior trauma, aggravated by repetitive weight-bearing activity, and relieved with rest and (2) a confirmatory radiograph and/or a nuclear bone scan at a site consistent with clinical presentation. A positive radiograph was defined as the presence of periosteal reaction, endosteal callous formation, and/or a fracture line in otherwise normal bone. A positive bone scan was defined as the presence of a 3× to 4× intensity localized fusiform uptake at the site of pain.

During the study it was apparent that some degree of under-reporting was likely and that some stress fractures were being missed with the passive follow-up. To get some measure of the degree of under-reporting, a sample of recruits was actively followed. For a period of 2 months toward the latter part of the study, all recruits (whether or not they were study participants) were returned to the Sports Medicine Clinic at the end of the 12 week training period. These subjects were given a questionnaire with questions regarding symptoms of stress-related injury. Any recruit with positive symptoms was examined, and any suspected fractures were confirmed using the preceding criteria.

Statistical analysis

Comparisons of anthropometric and bone structural data between normal subjects and those with stress fractures were carried out using two-tailed t-tests. Because of the greatly unequal sample sizes of the two groups (see below), separate (rather than pooled) variance t-tests were employed. Linear regression analysis and analysis of covariance were also used to further test the associations between body size, bone structural properties, and stress fracture incidence using the SYSTAT program (SYSTAT Inc., Evanston, IL, U.S.A.). A probability level of 0.05 was accepted as significant in all analyses.

RESULTS

A total of 23 of the 626 recruits (3.7%) presented with 27 lower-extremity stress fractures, consistent with previously reported rates for U.S. Marine recruits.[20] The most common site was the tibia (n = 11; 41%), followed by the metatarsals (n = 7; 26%), the femur (n = 5, 19%), and the tarsals (n = 4; 15%). Two subjects had fractures at two sites, and one subject had fractures at three sites. Because of the relatively small number of stress fracture cases, we did not attempt to break down the analysis by fracture site, i.e., all fracture cases were pooled for subsequent analysis. An additional six stress fractures were diagnosed in the 240 recruits who were interviewed at the completion of basic training, only one of whom was in the scanned cohort (included in the 23 above). The active follow-up indicated that the actual stress fracture rate in this population is approximately 6.2%, or stated another way, the self-reported stress fracture rate under-reports the actual rate by approximately 40%. In addition to stress fractures, an additional 16 recruits presented with shin splints, periostitis, or stress reactions. Because these subjects did not meet the strict criteria for stress fractures, they were neither included as fracture cases nor as normals for statistical comparisons (i.e., leaving 587 normal subjects).

Average values of age and anthropometric characteristics of normal subjects and those with stress fractures are shown in Table 2. Age is not significantly different between the two groups. Except for femur length, bi-iliac breadth, and bicondylar breadth, all anthropometric dimensions are significantly smaller in stress fracture subjects than in normals. The average difference in body weight between the two groups is almost 11%. Fracture cases are also relatively lighter for their height, as indicated by a significantly lower

TABLE 2. AVERAGE VALUES OF AGE AND ANTHROPOMETRIC DIMENSIONS IN SUBJECTS WITH STRESS FRACTURES AND NORMALS

	Fracture cases (n = 23)		*Normals (n = 587)*			
Measure	*Mean*	*SE*	*Mean*	*SE*	*% Difference**	*Significance*[†]
Age (years)	19.39	0.41	19.28	0.07	0.6%	(0.8)
Weight (kg)	67.29	2.41	75.42	0.46	−10.8%	0.0006
Height (cm)	171.41	1.19	175.04	0.27	−2.1%	0.01
BMI (kg/m^2)	22.85	0.70	24.59	0.13	−7.1%	0.01
Neck girth (cm)	37.48	0.40	38.59	0.09	−2.9%	0.02
Waist girth (cm)	81.31	2.05	85.11	0.36	−4.5%	0.04
Thigh girth (cm)	51.69	0.91	54.49	0.19	−5.1%	0.005
Calf girth (cm)	35.87	0.70	37.29	0.13	−3.8%	0.03
Tibia length (cm)	39.36	0.46	40.76	0.10	−3.4%	0.008
Femur length (cm)	51.32	0.56	52.17	0.12	−1.6%	(0.2)
Bi-iliac breadth (cm)	27.56	0.54	28.46	0.10	−3.2%	(0.08)
Femur bicondylar breadth (cm)	10.15	0.14	10.46	0.03	−3.0%	(0.06)

BMI, body mass index; SE, standard error of the mean. A probability level of 0.05 was accepted as significant in all analyses.

* [(fracture-normal)/normal] × 100.

† Values in parentheses not significant $p = 0.05$ level, two-tailed t-test.

TABLE 3. MEAN AND RANGE OF COEFFICIENTS OF VARIATION IN CROSS-SECTIONAL PROPERTIES AND BMD MEASURED BY DXA SCANNING FIVE TIMES IN EACH OF 10 MALE MARINE CORPS RECRUITS

	Mean % CV (range)		
Measurement	*Femur*	*Tibia*	*Fibula*
Cross-sectional area	1.04% (0.71–1.49%)	1.44% (1.07–2.06%)	2.55% (1.63–3.96%)
Cross-sectional moment of inertia	1.92% (1.10–2.75%)	3.02% (2.14–6.57%)	4.91% (2.90–6.36%)
Bone width	0.91% (0.31–1.87%)	1.45% (0.98–2.09%)	2.24% (1.50–2.97%)
BMD	1.35% (0.82–2.07%)	2.03% (1.57–2.43%)	2.26% (1.27–3.56%)

BMI. Average between-group differences in other body dimensions range between 2 and 5%.

Scan reproducibility for cross-sectional properties at the three measurement locations was determined as the percent coefficient of variation over the five repeated scans then averaged over the 10 subjects (Table 3). Theoretically one would expect that precision would be highest with the largest bone since larger bones have more individual pixel values (samples) incorporated in the measurement. This is the case (Table 3), where precision is greatest with the femur and worst with the fibula. Except for the cross-sectional moment of inertia of the fibula, mean CVs are similar for geometric properties and BMD ranging between approximately 1 and 3%.

Table 4 compares cross-sectional and bone mineral properties of the tibia, fibula, and femur between fracture cases and normals. Except for the fibula cross-sectional moment of inertia, section modulus, and width, all properties are significantly smaller in fracture subjects.

Because body size is also smaller in stress fracture subjects (Table 3), and all bone structural properties are significantly correlated with body size (correlations with body weight range from $r = 0.26$ for fibular width to $r = 0.72$ for femoral cross-sectional area), it is of interest to examine whether the structural properties remain smaller in fracture cases when variation in body size is controlled. Body weight was chosen as the best index of body size, since it has the highest and most consistent correlations with the other anthropometric variables (correlations range between $r = 0.36$ and $r = 0.86$). Also, theoretically, variation in body weight itself should constitute a considerable proportion of the variability in mechanical loading of the lower limb. An analysis of covariance was used to correct for differences in the average body weight between groups and to produce adjusted least squares means for cross-sectional geometric properties and femoral bicondylar breadth. (BMD was not included in this analysis, since it is already "standardized" by bone width, another "size" measure.)

TABLE 4. CROSS-SECTIONAL MEASUREMENTS AND BMD VALUES FOR FRACTURE CASES AND NORMALS

	Fracture cases (n = 23)		*Normals* (n = 587)			
Measurement	*Mean*	*SE*	*Mean*	*SE*	*% Difference**	*Significance*†
Tibia CSA (mm^2)	296.4	8.6	332.5	1.8	10.9%	0.0001
Tibia CSMI (mm^4)	10,419	651	13,112	148	20.5%	0.0004
Tibia width (mm)	20.5	0.3	21.8	0.1	6.0%	0.0005
Tibia Z (mm^3)	1,004	46	1,190	10	15.6%	0.0002
Tibia BMD (g/cm^2)	1.52	0.03	1.61	0.01	5.6%	0.006
Fibula CSA (mm^2)	86.6	3.3	96.3	0.7	10.1%	0.007
Fibula CSMI (mm^4)	901	77	1,072	20	16.0%	(0.09)
Fibula Width (mm)	11.4	0.3	11.8	0.1	3.4%	(0.3)
Fibula Z (mm^3)	153.6	8.8	176.4	2.1	12.9%	0.035
Fibula BMD (g/cm^2)	1.27	0.04	1.34	0.01	5.2%	(0.2)
Femur CSA (mm^2)	487.3	13.5	536.5	2.8	9.2%	0.0008
Femur CSMI (mm^4)	22,128	1,234	27,392	300	19.2%	0.0006
Femur width (mm)	23.5	0.4	24.8	0.1	5.2%	0.0008
Femur Z (mm^3)	1,863	75	2,177	413	14.4%	0.0004
Femur BMD (g/cm^2)	2.19	0.04	2.28	0.01	3.9%	0.03

CSA, CSMI, and Z, stand for cross-sectional area, moment of inertia, and modulus, respectively. A probability level of 0.05 was accepted as significant in all analyses; SE refers to the standard error of the mean.
* [(fracture-normal)/normal] · 100.
† Values in parentheses not significant at $p = 0.05$ level, two-tailed t-test.

TABLE 5. BONE STRUCTURAL MEASUREMENTS ADJUSTED FOR BODY WEIGHT IN FRACTURE CASES AND CONTROLS

	Fracture cases (n = 23)		*Normals* (n = 587)			
Measurement	*Mean*	*SE*	*Mean*	*SE*	*% Difference**	*Significance*†
Tibia CSA (mm^2)	316.7	6.7	331.8	1.3	−4.6%	0.03
Tibia CSMI (mm^4)	11,926	603	13,053	118	−8.6%	(0.07)
Tibia Z (mm^3)	1,109	38	1,186	7.5	−6.5%	0.05
Tibia width (mm)	21.04	0.31	21.72	0.06	−3.1%	0.03
Fibula CSA (mm^2)	92.1	3.2	96.1	0.6	−4.2%	(0.2)
Fibula CSMI (mm^4)	1,024	92	1,067	18	−4.0%	(0.7)
Fibula Z (mm^3)	168	10	176	2	−4.5%	(0.5)
Fibula width (mm)	11.68	0.32	11.74	0.06	−0.5%	(0.8)
Femur CSA (mm^2)	521.9	10.1	535.1	2.0	−2.5%	(0.2)
Femur CSMI (mm^4)	25,596	1,115	27,256	219	−6.1%	(0.2)
Femur Z (mm^3)	2,069	61	2,169	12	−4.6%	(0.1)
Femur width (mm)	24.24	0.33	25.80	0.06	−6.0%	(0.1)
Femur bicondylar breadth (cm)	10.47	0.13	10.44	0.02	0.3%	(0.8)

CSA, CSMI, and Z stand for cross-sectional area, moment of inertia, and modulus, respectively. A probability level of 0.05 was accepted as significant in all analyses; SE refers to the standard error of the mean.
* [(fracture-normal)/normal] · 100.
† Values in parentheses not significant at $p = 0.05$ level, two-tailed t-test.

Table 5 presents the comparison of body weight adjusted properties in fracture and normal groups. When the body weight is controlled, most bone structural properties are not significantly different between groups, although all properties except femoral bicondylar breadth are still lower in fracture cases. The only significant differences are for tibia cross-sectional area, section modulus, and width. Thus, for these properties, fracture cases are both absolutely and relatively small compared with normals.

This point is illustrated in Figs. 1 and 2, which show plots of tibia cross-sectional area and section modulus against body weight in normals and fracture cases. In general, stress

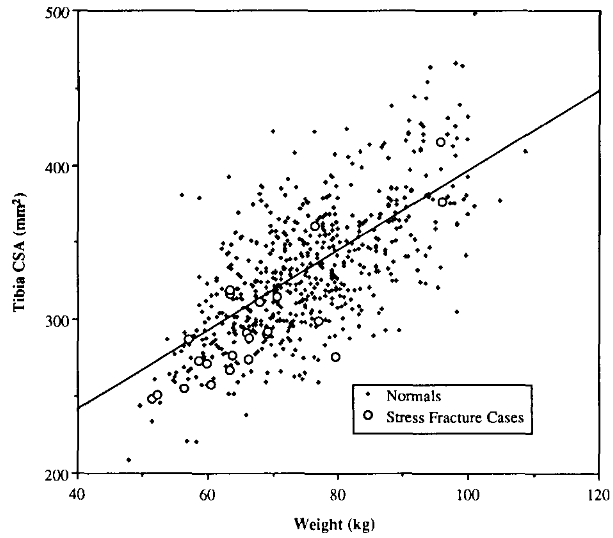

FIG. 1. CSA of the distal third of the tibia as a function of body weight, with linear regression through normals ($r = 0.66, p < 0.001$). Fracture subjects are generally low in body weight but also have low CSAs for their body weight.

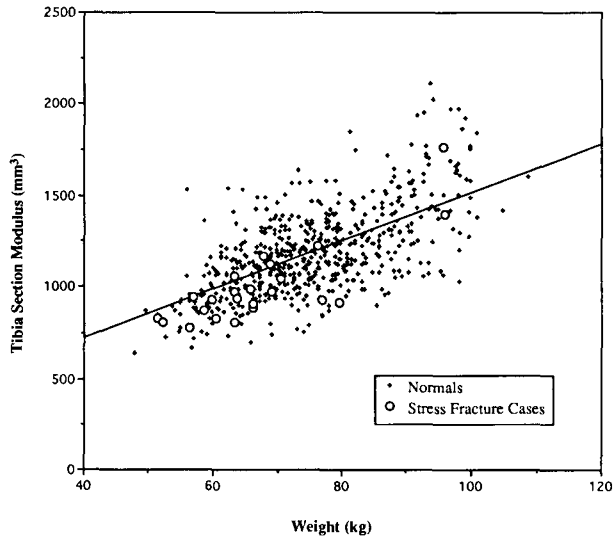

FIG. 2. Section modulus (Z) of the distal third of the tibia as a function of body weight, with linear regression through normals ($r = 0.63, p < 0.001$). Like CSA, Zs of fracture subjects are generally low for their body weight.

fracture subjects are small in body size compared with normals (78% fall below the median weight of normals, and 61% fall in the lower quartile of normals), but their tibial cross-sectional properties are also, on average, small relative to normal subjects of similar body weight. This is not simply a function of a relatively smaller skeletal size in fracture subjects. Figure 3 is a similar plot of femoral bicondylar breadth against body mass and shows that joint size relative to body size is not smaller in fracture subjects (see also Table 5). The contrast in diaphyseal-to-joint proportions in fracture cases and normals is illustrated in Fig. 4, where the femoral midshaft breadth is regressed against the bicondylar breadth. The ratio of shaft breadth to bicondylar breadth is significantly smaller in fracture subjects ($p < 0.026$, two-tailed t-test).

DISCUSSION

Previous studies using conventional bone mineral analysis methods to correlate BMD with stress fracture incidence have shown mixed results, with some studies demonstrating significant differences in BMD between normals and stress fracture cases and some not.[21–23] Although bone width has been measured absorptiometrically by others,[22] the present study represents the first use of the DXA method for the investigation of bone cross-sectional properties in a prospective study of factors predisposing to stress fracture. In this study, both small body size and narrow long bone diaphyses relative to body size were predisposing factors for developing lower limb stress fractures during the 12 week basic training program. Smaller body size may be a risk factor in part because of common training requirements involving carrying of similar weight packs and other equipment regardless of recruit body weight. For example, in one study of Army cadets at the U.S. Military Academy,[24] males and females were required to carry equal weight packs (16 kg) during training, despite almost certain gender differences in body size which may in part have contributed to a much higher female stress fracture rate in this sample (see also Pester and Smith[25]). The approximate combined weight of packs and equipment carried by recruits in our sample during some parts of their training regimen totaled about 23 kg. This represents a 34% increase in body weight for the stress fracture group but a lower 30% increase in body weight for the normal subjects. It is also possible that the relative thinness (lower BMI) of the fracture group is indicative of relatively lower muscle mass and/or poorer physical conditioning prior to training. Our anthropometric girth measurements do not allow a distinction between muscle and fat soft tissue components. However, it may be possible to use the DXA data to derive this information at the lower limb scanning sites. We are currently exploring this possibility using also the physical conditioning records obtained at the time of induction of each recruit.

The smaller cross-sectional areas and section moduli of fracture subjects relative to body mass, statistically significant in the tibia, are also predisposing factors to increased stress fracture risk. The sudden onset of higher levels of activity and thus increased bone strain in individuals with relatively weaker bones, without time for adaptive cortical remodeling, would be expected to lead to a higher rate of bone microdamage resulting in stress fractures.[26,27] The apparent restriction of smaller dimensions in the fracture group to long bone diaphyses, but not joint size, is interesting in that it suggests a specificity in the structural "deficit" present in this group. There is evidence that joint size is more genetically determined and less environmentally plastic than diaphyseal cross-sectional dimensions.[9,28,29] Thus, the fact that diaphyses but not joints of fracture subjects are

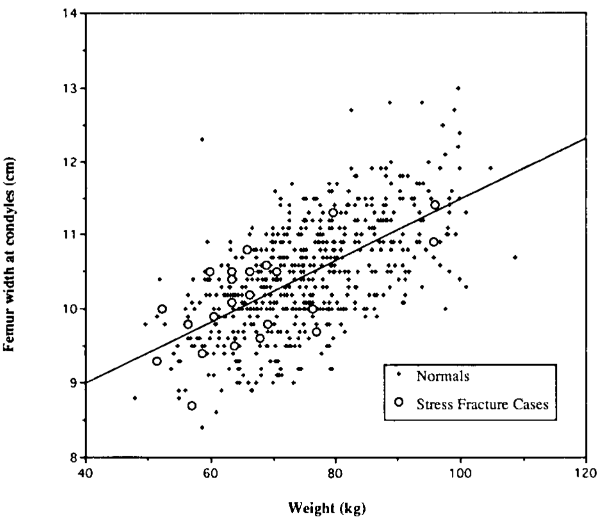

FIG. 3. Breadth of the femoral condyles as a function of body weight with linear regression through normals ($r = 0.61$, $p < 0.001$). Relative knee joint size is not dissimilar between stress fracture cases and normals.

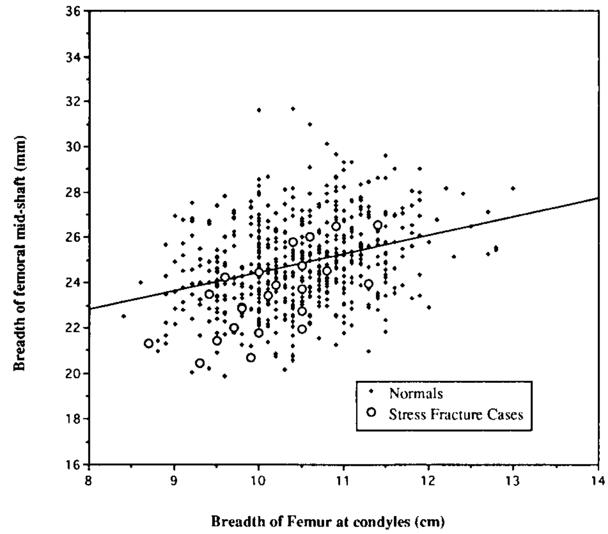

FIG. 4. Breadth of the femur at midshaft as a function of breadth of the femoral condyles with linear regression through normals ($r = 0.32$, $p < 0.001$). Fracture subjects have relatively small bone shaft diameters for their joint size.

smaller suggests an environmental explanation for their increased fracture risk, namely that their bones had not been loaded sufficiently prior to induction to develop adequately strong cortices. If such subjects can be identified prior to training, greater emphasis on increased physical activity well before induction may alleviate this problem.

The individual data scatters shown in Figs. 1 and 2 demonstrate that not all individuals who developed stress fractures in our sample followed the typical pattern of small body size and average to small diaphyseal dimensions relative to body size. Two of the 23 stress fracture subjects were among the heaviest individuals in the total sample. In contrast to the other stress fracture subjects, they were also among the highest in BMI (weight/height^2) in the entire study sample. It is possible, therefore, that, for some individuals, being relatively overweight could be a contributing factor for developing stress fractures under conditions of increased mechanical loading. More information on fat and muscle soft tissue components might again be useful in evaluating such individuals. Perhaps a relatively small muscle mass in these heavy individuals may fatigue sooner, resulting in greater bone flexing under repetitive load.

In a previous study of Israeli infantry recruits,[13,14] no relationship between body size (weight and height) and stress fracture risk was found. When our bone structural variables are adjusted for differences in body weight (Table 5), the average difference in the tibia cross-sectional moment of inertia between fracture and normal subjects is −8.6%, similar to the average (non–weight-adjusted) difference between groups in the tibia cross-sectional moment of inertia reported by these authors (−7.8%, mean of cross-sectional moment of inertia measured about the anteroposterior axis at two locations in the distal half of the tibia). Thus, relative to body weight, the average reduction in mediolateral bending rigidity of the tibia in stress fracture subjects appears to be similar in the two studies. Why body weight itself should be such an important factor in U.S. Marine but not in Israeli infantry recruits is difficult to say. It is possible that the physical characteristics or the level of prior physical conditioning of military trainees differ significantly in the two countries. Another possible contributing factor is differences in the case definition for stress fracture used in the Israeli and present studies. In the Israeli study all subjects were examined by both radiography and bone scintigraphy, but only 20% of the stress fractures confirmed by scintigraphy were radiographically positive.[3] Indeed, that study indicated that 69% of the femoral stress fractures were asymptomatic and thus would not have been reported using the methods of the current study. Jones et al.[20] suggest that the Israeli stress fracture rate when corrected for the difference in case definition would be approximately 6.2%. Interestingly, this is equivalent to the level of stress fractures in the current study when self-referral is supplemented by active surveillance for symptoms.

In summary, U.S. Marine Corps male recruits who developed lower limb stress fractures during an initial 12 week training period were, on average, significantly smaller in body size and had significantly weaker tibial diaphyses relative to body weight than did recruits who did not develop stress fractures. Bone structural data derived from DXA provides important new information that may be useful in the identification of subjects at higher risk for stress fractures under intense physical training conditions. The method for computation of geometry from DXA data should also be more accurate than radiographic methods employed in other studies[14] which entail assumptions of cross-sectional shape and manual measurements of cortical thickness.

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APPENDIX: DERIVATION OF CROSS-SECTIONAL GEOMETRY FROM BONE MASS DATA

DXA data were acquired in a series of 10 scan lines traversing the leg at the two scan locations. After calibration, a measurement of bone mass in grams per square centimeter is computed from stored data corresponding to each point (pixel value) on each scan line. Pixel values measured over bone represent the total bone mass summed along a line extending through the patient at that point, and those over soft tissue that contain no bone mass are approximately zero. We have written programs (in cooperation with manufacturers) to extract individual pixel values from the raw scan data. Cross-sectional properties are computed from profiles of bone mass values in scan lines traversing the bone. Since scan lines through the lower leg traverse both tibia and fibula, two profiles are obtained and treated individually. Bone widths are calculated as the distance between profile margins after correction for image blurring. The cross-sectional area (CSA) at the location of the mass profile is obtained by the summation

$$CSA = \frac{\Delta L}{\rho_b} \sum_{i=1}^{N} m_{bi} \tag{1A}$$

where ΔL is the pixel spacing along the profile line, ρ_b is the density of cortical bone (1.88 g/cm^3 is assumed), m_b is the bone mass pixel value, and N is the number of pixels in the profile. Note that the cross-sectional area defined here excludes trabecular and cellular voids (excluded soft tissue is not mechanically relevant at physiologic strain rates[30]). To determine the cross-sectional moment of inertia (CSMI), the location of the neutral axis is first obtained as the coordinate of the centroid (x_c) of the bone mass distribution in the profile. The CSMI is then calculated as

$$CSMI = \frac{\Delta L}{\rho_b} \sum_{i-1}^{N} (x_i - x_c)^2 m_{bi} \tag{2A}$$

(Note that since DXA scans in this study were taken in the frontal plane, CSMI's are relevant only for mediolateral bending; an assessment of anteroposterior bending would require a second scan in the lateral projection.) To maximize statistical precision, the CSA, CSMI, and bone widths were averaged over the 10 scan lines at each location.

REFERENCES

1. Gilbert RS, Johnson HA 1966 Stress fractures in military recruits—A review of twelve years experience. Military Med **131**:716–721.
2. Greaney RB, Gerber FH, Laughlin JP, Kmet JP, et al. 1983 Distribution and natural history of stress fractures in U.S. Marine Corps Recruits. Radiology **146**:339–346.
3. Milgrom C, Giladi M, Stein M, Kashtan H, Margulies JY, Chisin R, Steinberg R, Aharonson Z 1985 Stress fracture in Military Recruits: A prospective study showing unusually high incidence. J Bone Joint Surg **67-B**:732–735.
4. Kimsey CD 1993 The epidemiology of lower extremity injuries in United States Marine Corps recruits. Ph.D. dissertation, University of South Carolina.
5. Lanyon LE, Hampson WGJ, Goodship AE, Shah JS 1975 Bone deformation recorded in vivo from strain gages attached to the human tibial shaft. Acta Orthop Scand **46**:256–268.
6. Frankel VH, Burstein AH 1965 Load capacity of tubular bone. In: Kenedi RM (ed.) Biomechanics and Related Bioengineering Topics. Pergamon Press, pp. 381–396.
7. Martens M, Van Audekerke R, de Meester P, Mulier JC 1981 The geometrical properties of human femur and tibia and their importance for the mechanical behavior of these bone structures. Arch Orthop Traumatic Surg **98**:113–120.
8. Ruff CB, Hayes WC 1988 Sex differences in age related remodeling of the femur and tibia. J Orthop Res **6**:886–896.
9. Ruff CB, Scott WW, Liu A, Y-C 1991 Articular and diaphyseal remodeling of the proximal femur with changes in body mass in adults. Am J Phys Anthrop **86**:397–413.
10. Ruff CB, Walker A, Trinkaus E 1993 Postcranial robusticity in Homo I: Temporal trends and mechanical interpretation. Am J Phys Anthopol **91**:21–53.
11. Beck TJ, Ruff CB, Scott WW, Plato CC, Tobin JD, Quan CA 1992 Sex differences in geometry of the femoral neck with aging: A structural analysis of bone mineral data. Calcif Tissue Int **50**:24–29.
12. Cowan SC 1983 The mechanical and stress adaptive properties of bone. Ann Biomed Eng **11**:263–295.
13. Giladi M, C. M, Simpkin A, Stein M, Kashtan H, Margulies J, Rand N, Chisin R, Steinberg R, Aharonson R, Kedem R, Frankel VH 1987 Stress fractures and tibial bone width as risk factor. J Bone Joint Surg **69-B**:326–329.
14. Milgrom C, Giladi M, Simkin A, Rand N, Kedem R, Kashtan H, Stein M, Gomori M 1989 The area moment of inertia of the tibia: a risk factor for stress fractures. J Biomech **22**:1243–1248.
15. Martin R, Burr D 1984 Non-invasive measurement of long bone cross-sectional moment of inertia by photon absorptiometry. J Biomech **17**:195–201.
16. Beck TJ, Ruff CB, Warden KE, Scott WW, Rao GU 1990 Predicting femoral neck strength from bone mineral data: A structural approach. Invest Radiol **25**:6–18.
17. Jakob R, Haertal M, Stussi E 1980 Tibial torsion calculated by computerized tomography and compared to other methods of measurement. J Bone Joint Surg **62B**:238–242.
18. Beck TJ, Ruff CB, Bissessur K 1993 Age-related changes in female femoral neck geometry: Implications for bone strength. Calcif Tissue Int (Suppl 1):S41–S46.
19. AAOS 1991 Orthopaedic ICD9-CM Expanded, 2nd Ed. American Academy Orthopaedic Surgeons.
20. Jones BH, Harris JM, Vinh TN, Rubin CR 1989 Exercise-induced stress fractures and stress reactions of bone: epidemi-

ology, etiology, and classification. In: Exercise and Sports Sciences Reviews, Vol 17. Williams & Wilkins, pp. 379–422.
21. Giladi M, Milgrom C, Simkin A, Danon Y 1991 Stress fractures: Identifiable risk factors. Am J Sports Med **19**:647–652.
22. Margulies JY, Simkin A, Leichter I, Bivas A, Steinberg R, Giladi M, Stein M, Kashtan H, Milgrom C 1986 Effect of intense physical activity on the bone-mineral content in the lower limbs of young adults. J Bone Joint Surg **68**:1090–1093.
23. Pouilles JM, Bernard J, Tremollieres F, Louvet JP, Ribot C 1989 Femoral bone density in young male adults with stress fractures. Bone **10**:105–108.
24. Protzman RR, Griffis CG 1977 Stress fractures in men and women undergoing military training. J Bone Joint Surg **59A**: 825.
25. Pester S, Smith PC 1992 Stress fractures in the lower extremities of soldiers in basic training. Orthop Rev **21**:297–303.
26. Carter DR 1982 The relationship between in vivo strains and cortical bone remodeling. [Review]. Crit Rev Biomed Eng **8**:1–28.
27. Lanyon LE, Goodship AE, Pye CJ, MacFie JH 1982 Mechanically adaptive bone remodelling. J Biomech **15**:141–154.
28. Ruff CB, Walker A, Trinkaus E 1994 Postcranial robusticity in Homo, III: Ontogeny. Am J Phys Anthrop **93**:35–54.
29. Trinkaus E, Churchill SE, Ruff CB 1994 Postcranial robusticity in Homo, II: Humeral bilateral asymmetry and bone plasticity. Am J Phys Anthropol **93**:1–34.
30. Martin R 1991 Determinants of the mechanical properties of bones. J Biomech **24**, (Suppl 1):79–88.

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