ELSEVIER

Bone Vol. 27, No. 2
August 2000:319–326

Bone-Muscle Strength Indices for the Human Lower Leg

J. RITTWEGER,[1,2] G. BELLER,[2] J. EHRIG,[2] C. JUNG,[2] U. KOCH,[2] J. RAMOLLA,[2] F. SCHMIDT,[2] D. NEWITT,[3] S. MAJUMDAR,[3] H. SCHIESSL,[4] and D. FELSENBERG[2]

[1]*Institute of Physiology, Freie Universität Berlin, Berlin, Germany*
[2]*Osteoporosis Research Group, Freie Universität Berlin and University Hospital Benjamin Franklin, Berlin, Germany*
[3]*Magnetic Resonance Science Center, University of California at San Francisco, San Francisco, CA, USA*
[4]*Stratec Medizintechnik, Pforzheim, Germany*

This cross-sectional study is based on images from the lower leg as assessed by peripheral quantitative computer tomography (pQCT). Measurements were performed in 39 female and 38 male control subjects and 15 female professional volleyball players, all between 18 and 30 years of age. The images were obtained at shank levels of 4%, 14%, 33%, and 66% from the distal end. Bone and muscle cross-sectional areas, and the bones' density-weighted area moment of resistance and of inertia were assessed. From these, muscle-bone strength indices (MBSIs) were developed for compression (CI = 100 · bone area/muscle area) and bending (BI = 100 · bone area moment of resistance/muscle area/tibia length). Significant correlations between muscle cross-sectional area and bone were found at all section levels investigated. The strongest correlation for compression was observed in the sections at 14% (correlation coefficient r = 0.74), where 4.10 ± 0.46 cm^2 bone, on average, was related to 100 cm^2 muscle. The compression index (CI) at the 14% level was independent of the tibia length. Interestingly, the 15 athletes had significantly greater CIs than the control subjects. This is most probably due to the greater tension development in the athletes. The highest correlation for bending was for anteroposterior bending at 33% of tibia length (r = 0.81), where the area moment of resistance, R, was on, average, 4.21 ± 0.54 cm^3/100 cm^2 muscle/m tibia length. Analysis of the bones' area moment of inertia showed that buckling is a possible cause of bending at the 33% and 66% levels, but not at the 14% level. No gender differences in MBSI were found. Likewise, age was without significant effect. The data show that bone architecture depends critically on muscle cross section and tension development. Moreover, bone geometry (e.g., the tibia length) influences the geometrical distribution of bone mineral, as it was found that long bones adapted to the same compressive strength are wider than short ones. We conclude that MBSIs offer a powerful diagnostic tool for bone disorders and may contribute to improving the treatment of bone metabolic and other diseases. (Bone 27:319–326; 2000)

Key Words: Anthropometry; Biomechanics; Bone physiology; Muscle physiology; Osteoporosis; Peripheral quantitative computer tomography (pQCT).

Introduction

Bones adapt their strength to the mechanical forces they undergo.[2,18] This is accomplished by either the modeling process,[12] which can add net bone, or by remodeling, which can remove bone.[13] As a whole, this system works as a "mechanostat," keeping the strains exerted by external forces within certain limits.[11] The external forces arise essentially from muscle contractions. Consequently, exercise-typical, site-specific relations between muscle force and bone geometry have been demonstrated.[20,31]

With dual-energy X-ray absorptiometry (DXA), body composition can be assessed and broken down into: (i) bone mineral content (BMC); (ii) fat body mass (FBM); and (iii) lean body mass (LBM), half of which is closely correlated with muscle mass.[19] Strong correlations have been observed between LBM and total body BMC.[25] In fact, BMC is influenced by an order of magnitude stronger by LBM than by FBM or by height.[6]

Considered over the whole body, BMC amounts to approximately 5% of LBM, except in premenopausal women, who store 17%–29% more bone mineral per LBM than prepubertal children, postpubertal men, or postmenopausal women.[6,25] It has been hypothesized that this "surplus" of bone mineral is stored at locations where it is not needed mechanically.[7]

Given the strong interrelationship between muscle force and bone strength, diagnostic indices that quantify these relations may turn out to be valuable tools in the diagnosis of osteoporosis and other bone disorders. This approach has been perceived and assessed in several studies.[7,22,27] Lately, the term "bone-muscle strength index" (BMSI) has been coined for such quantitative measures (H. M. Frost, personal communication, 1999).

Physiologically, bones are loaded by compression as well as by bending.[3,4] In the lower leg, bending may occur by the horizontal component of the soleus muscle fibers, by eccentric muscle pull, or through buckling under mere compression. Separate BMSIs should therefore be developed for the two strain mechanisms. With adequate tools, peak muscle force can be assessed through a limb's torque during maximum voluntary contraction and analysis of the muscle levers. Likewise, the potential force development of muscles could also (and may be more objectively) be quantified by their cross-sectional area (CSA).[21] Respectively, the bending moment that a certain mus-

Address for correspondence and reprints: Dr. Jörn Rittweger, Institut für Physiologie, Arnimallee 22, 14195 Berlin, Germany. E-mail: ritmus@zedat.fu-berlin.de

Bone Vol. 27, No. 2
August 2000:319–326

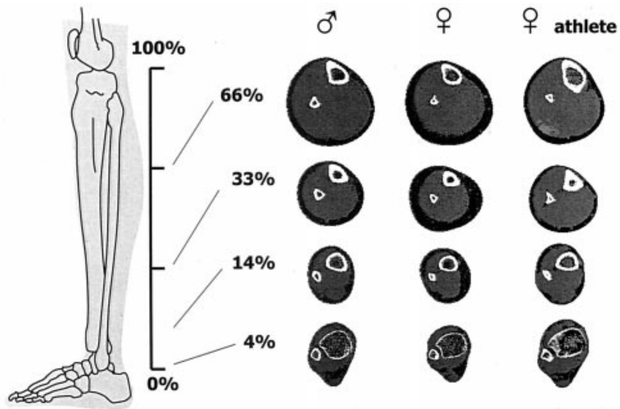


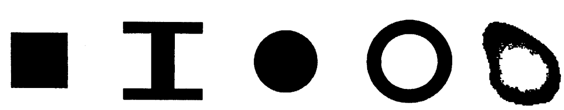
Figure 1. Overview of pQCT sections obtained. Sections were obtained at 4%, 14%, 33%, and 66% of the tibia length (between the medial knee joint cleft and the medial malleolus, both palpated manually). The men in group 1 had a significantly larger bone cross-sectional area (CSA) and muscle CSA than the women. Moreover, the women's subcutaneous fat tissue was greater. In this respect, the female athletes (group 2) resemble normal men. Compared with their muscle CSA, they seem to have a larger bone CSA.

cle exerts could be identified by the product of CSA and the length of the lever.

The material constants of bone are largely untouched by age, gender, and species,[1,32] but bone as a mechanical structure is crucially dependent on its apparent density. Thus, good prediction of a bone's ultimate strength and stiffness is possible by quantitative computer tomography (QCT).[5,30] The appropriate resistance measures for compression and for bending are the bone CSA and the area moment of resistance, respectively. For an illustration of the area moment of resistance, see **Figure 2**.

The present investigation develops BMSI for compression and bending at the human lower leg. To do so, the following hypotheses are tested:

1. *H*1: There are significant muscle and bone relationships for indices of compression and bending at the lower leg.
2. *H*2: The muscle-bone relationship at the lower leg will depict a gender difference, revealing women to have more bone mineral per muscle tissue than men.
3. *H*3: A supposed greater tension production in athletes will affect the muscle-bone relationship as assessed by imaging techniques.



	square	double-T	circle	ring	tibia
A	3,1 cm²	3,1 cm²	3,1 cm²	3,1 cm²	3,1 cm²
R_y	0,91 cm³	1,61 cm³	1,11 cm³	1,48 cm³	1,45 cm³
I_y	0,79 cm⁴	2,07 cm⁴	1,24 cm⁴	1,97 cm⁴	2,03 cm⁴
R_x	0,91 cm³	1,93 cm³	1,11 cm³	1,48 cm³	1,56 cm³
I_x	0,79 cm⁴	2,40 cm⁴	1,24 cm⁴	1,97 cm⁴	2,23 cm⁴

Figure 2. Illustration of the area moment of resistance (*R*) and the area moment of inertia (*I*). Given are four beams with simple geometric forms, and the human tibia, all five with identical cross-sectional areas (*A*). The greater the *R*, the greater the beam or bone resistance to bending. The greater the *I*, the greater the resistance to buckling. *I* and *R* vary with the direction. Given are the values for *x* (anteroposterior) and *y* (lateral) flexion. It can be seen that, the further the material is from the structure's center, the greater the *I* and *R*. Thus, material eccentricity provides an idea of a structure's relative adaptation to buckling and bending.

Table 1. Anthropometric data (mean ±SD)

	N	Age (years)	Weight (kg)	Height (cm)	L_{Tibia} (cm)
Group 1					
Women	39	24.9 ± 3.5	61.3 ± 7.1	168.6 ± 5.1	37.2 ± 2.7
Men	38	24.9 ± 3.6	73.0 ± 8.8	180.7 ± 5.7	41.3 ± 3.4
Group 2					
Women	15	21.3 ± 3.8	69.7 ± 4.0	182.9 ± 4.9	42.0 ± 2.0

Materials and Methods

Setup and Subjects

Two different groups of subjects were investigated. Group 1 was comprised of 78 female and male whites native to Berlin, between 20 and 30 years of age. Group 2 consisted of 15 female whites who were professional volleyball players in Berlin, all of whom were competing at the highest national or international level. The biometric data of the three groups are given in **Table 1**. Groups 1 and 2 underwent analysis by peripheral quantitative computer tomography (pQCT). The investigations were approved by the local ethics committees. All subjects gave written informed consent before inclusion in the study.

All subjects were in good health. This was confirmed by: (i) clinical history; (ii) physical examination; and (iii) blood sample. All women had had a regular menses during the last 24 months before the investigation. No female subject in group 1 had been pregnant prior to inclusion in the study, nor had any taken an oral contraceptive during the last 6 months. In group 2, three women were on oral contraceptives; two women had been pregnant prior to the study, in both cases >2 years previously.

The length of the tibia (L_{Tibia}) was assessed as the distance between the medial knee joint cleft and the medial malleolus (both palpated manually). The repeatability of this measuring procedure was tested beforehand; it yielded maximal errors of 0.3 cm (five subjects, ten repetitions each). Between these two points, the level of 66% L_{Tibia} from the ankle was marked manually.

pQCT Analysis

pQCT images were obtained using an XCT-2000 apparatus (Stratec Medizintechnik, Pforzheim, Germany). The tibiotalar joint cleft was identified by scout viewing, thus giving the 4%, 14%, and 33% levels. Image analysis was performed with integrated software, version 5.4. Using this software, measures of bone mineral density are normalized to the density of fat tissue. Accordingly, the density of muscle tissue is about 70 mg/cm^3, whereas completely mineralized bone has a density of about 1200 g/cm^3. Images were taken with a voxel size of 0.4 mm (parameter voxel size in the integrated software) in the transverse direction and 2.4 mm in the longitudinal (X-ray beam width).

In all measurements, the angle between the foot and tibia was adjusted to 120°. The muscle CSA A_{Muscle} was assessed in the 66% section, because this is the region of largest outer calf diameter. Previous experiments have shown that A_{Muscle}, assessed at 60%, 63%, 69%, and 72%, was 1.025-, 1.021-, 0.961-, and 0.911-fold of the 66% CSA, respectively. Around the 66% section level, A_{Muscle} thus decreased by roughly 1% per percent section height. The corresponding relations in A_{Bone} were 1.011-, 1.015-, 0.990-, and 0.985-fold, and thus only about 0.5% per percent section level.

In the pQCT images, a region of interest (ROI) was defined within the subcutaneous tissue, but outside the muscle. Within this ROI, the total area of the muscle was determined with the threshold set at 35 mg/cm^3 (contour mode 1); that is, midway between the typical densities of muscle tissue (70 mg/cm^3) and fat tissue (0 mg/cm^3). Next, the total areas of the fibula and the tibia were assessed at contour mode 1 and the threshold set at 710 mg/cm^3. The areas were subtracted from the total muscle area and thus yielded A_{Muscle}.

For assessment of the bone CSA at the 4%, 14%, 33%, and 66% levels, the threshold for the outer edge was set at 710 mg/cm^3 (contour mode 1) and the bone mineral content per 1 cm slice was determined. The CSA of each bone was computed as A [cm^2] = BMC [g/cm]/1200 [g/cm^3]. For each section level, the areas of tibia and fibula were added as $A_{Bone} = A_{Tibia} + A_{Fibula}$.

A_{Bone} is a surrogate of the "bone cross-sectional area," which usually exceeds the area obtained by area segmentation (in the integrated software CRT_A). Results obtained with CRT_A are qualitatively comparable to those with A_{Bone}, and thus are not reported here. The excess of A_{Bone} over CRT_A amounted to 19.3%, 3.5%, and 6.2% at the 14%, 33%, and 66% section levels, respectively. In cancellous bone, A_{Bone} can be thought of as the bone area, if all trabeculae are packed tightly. The purpose for computing A_{Bone} through BMC is fourfold: (i) the method considers cancellous bone (present at 4%, and partly at the 14% level) and cortical bone (at 14%, 33%, and 66% levels), and thereby (ii) allows for comparison of the four section levels in this study and (iii) with other DXA-based studies, and (iv) is very simple.

The density-weighted area moment of resistance (R) in the x and y directions (termed SSIx and SSIy in the software) was assessed with the periosteal and endocortical thresholds set at 280 and 480 mg/cm^3, respectively. This yielded the variables R_X and R_Y, where R is the bone's moment of resistance, X represents bending in the anteroposterior direction, and Y denotes lateral bending (see Figure 2). The integrated software provides a density-weighted measure for R (termed SSI), which has the advantage that incompletely mineralized bone contributes less to the value—and therefore applied in this study. Values of tibia and fibula were added as $R_{Tibia} + R_{Fibula}$. Likewise, the density-weighted area moments of inertia, I_X and I_Y, were assessed as a measure of the bone's resistance to buckling. R_X, R_Y, I_X, and I_Y were assessed at the 14%, 33%, and 66% levels.

Last, the total bone area, defined as the bone's outer edge, was assessed at the 4% level as a measure of the joint area. Areas of tibia and fibula were summed as TA_{Bone}.

For all the aforementioned parameters, the short-term error (EST) was ascertained by two repeat measurements (x_1 and x_2) with a 2 week interval in between in 13 subjects. The mean of the two measurements, $\bar{x}$, was computed, and hence:

$$\mathrm{EST} = 100 \cdot \frac{\mathrm{Abs}(\bar{x} - x_1) + \mathrm{Abs}(\bar{x} - x_2)}{2 \cdot \bar{x}}$$

where Abs is the absolute value.

Data Analysis

A surrogate for the torque exerted by a muscle, "muscle bending moment" (MBM), was computed as $\mathrm{MBM} = A_{Muscle} \cdot L_{Tibia}$. The underlying assumptions are that the force increases linearly with the muscle CSA, and hence that the bending moment exerted on the bone increases with (i) the muscle CSA and (ii) with the length. It is assumed that the anatomical proportions of the musculature are size-independent. Because A_{Muscle} is in square centimeters, in order to understand MBM in (N/m) one should multiply parenthetically the tension developed by the muscle in (N/cm^2). Here, we measure MBM in (cm^2/m).

The compression indices (CIs) were computed as:

$$\mathrm{CI} = 100 \cdot \frac{A_{Bone}[\mathrm{cm}^2]}{A_{Muscle}[\mathrm{cm}^2]}$$

yielding the CIs for the different section levels, CI_{04}, CI_{14}, CI_{33}, and CI_{66}. For convenience, the CI values are given in percent; therefore, they denote: percent cm^2 bone per cm^2 muscle.

Except for the 4% level, MBSIs were computed in a similar manner, yielding the bending index (BI):

$$\mathrm{BI} = 100 \cdot \frac{R_{Bone}[\mathrm{cm}^2]}{\mathrm{MBM}[\mathrm{cm}^2/\mathrm{m}]}$$

Thus, the new variables, BI_{X14}, BI_{Y14}, BI_{X33}, BI_{Y33}, BI_{X66}, and BI_{Y66}, were obtained. Again, for convenience they are expressed in percent per meter (%/m).

Statistical analyses were conducted with the SPSS software in its PC version (v8.01). Before t-test or analysis of variance (ANOVA) was performed, variables were tested for normal distribution with the Kolmogorov–Smirnoff test, and for homogeneity of variances with Levene's test ($p > 0.25$). Whenever data depicted a normal distribution, differences between groups were checked by the t-test, or by one-way ANOVA with Bonferroni's correction for multiple comparisons. In all other cases, Wilcoxon or Friedman tests were applied. Correlation analysis was performed to obtain a Pearson correlation coefficient (r). Significance was assumed if $p < 0.05$.

Table 2. Short-term error (EST) for peripheral quantitative computer tomography (pQCT) measurements

	Level			
	4%	14%	33%	66%
A_{Muscle}	—	—	—	1.15%
TA_{Bone}	0.35%	—	—	—
A_{Bone}	0.33%	0.20%	0.17%	0.28%
R_{XBone}	—	1.30%	1.73%	1.24%
R_{YBone}	—	1.57%	1.58%	3.46%
I_{XBone}	—	0.95%	1.48%	2.55%
I_{YBone}	—	0.67%	1.92%	5.14%

EST was calculated as the deviation from the common mean in two assessments. Errors were greater in the higher moment parameters R and I.
KEY: A, angle; I, inertia; R, resistance; TA, total area.

Results

Group 2 was 3.5 years younger than group 1 ($p < 0.05$). Within group 1, men were significantly taller and heavier than women, and they had a longer tibia. The female athletes' (group 2) height, weight, and tibia length were indistinguishable from the men in group 1 (see Table 1).

The short-term error values (ESTs) of all variables, measured with pQCT, are given in **Table 2**. For A_{Bone}, EST was below 1%, and for A_{Muscle} it was 1.15%. In the higher moment parameters, errors were between 0.67% and 5.14%.

Bone Vol. 27, No. 2
August 2000:319–326

Table 3. Cross-sectional peripheral quantitative computer tomography (pQCT) measures

		A_{Bone} (cm^2)			
	A_{Muscle} (cm^2) at 66% level	4% level	14% level	33% level	66% level
Group 1					
Women	$67.8 \pm 7.6^{a,b}$	$3.50 \pm 0.49^{a,b}$	$2.76 \pm 0.23^{a,b}$	$3.55 \pm 0.32^{a,b}$	$3.86 \pm 0.31^{a,b}$
Men	83.1 ± 10.6	4.36 ± 0.52	3.37 ± 0.32	4.29 ± 0.42	4.57 ± 0.42
Group 2	74.1 ± 8.8	4.20 ± 0.37	3.1761 ± 0.25	4.06 ± 0.39	4.65 ± 0.48

Mean $\pm$ SD values of CSA (in cm^2).
[a]Different from group 2.
[b]Different from group 1 men.

Mean values $\pm$ SD of TA_{Bone} were 11.8 $\pm$ 1.68 cm^2 in women and 13.5 $\pm$ 1.59 cm^2 in men of group 1 ($p > 0.05$).

Compression

Muscle and bone cross-sectional areas are given in **Table 3**. Tests for differences were made within and among groups 1 and 2. The men in group 1 had significantly larger A_{Muscle} than women in group 1, but not larger than the female athletes of group 2. Likewise, A_{Bone} was significantly smaller in the women in group 1 than in the men, and also smaller than in the female athletes of group 2. This was true for all section levels.

In all subjects, A_{Bone} was lowest at the 14% level and greatest at the 66% level ($p < 0.05$; for means see Table 3). The values for the different levels were strongly correlated with each other, particularly the neighboring levels. The r^2 values ranged from between 0.71 and 0.83 (**Table 4**).

Significant correlations were found between A_{Muscle} and A_{Bone} at the 4%, 14%, 33%, and 66% levels. In group 1, r^2 was lowest for A_{Bone04} (0.44) and highest for A_{Bone33} (0.54, see Fig. 3A).

Due to the variation in A_{Bone}, CIs varied over the different section levels (see Fig. 4). Values ranged from between 4.09 (CI_{14} in women in group 1) and 6.33 (CI_{66} in group 2). No difference was found between the CIs of men and women within group 1 ($p > 0.25$), but CI_{04} and CI_{66} values in the athletes from group 2 were significantly higher than in group 1 ($p < 0.05$). Because there was no gender difference in group 1, the mean values over all subjects were computed (**Table 5**). No correlation between either of the CIs was found with age, height, body weight, or L_{Tibia} ($p > 0.1$).

The CI residuals (RSD_{CI}) were computed according to:

$$\mathrm{RSD}_{\mathrm{CI}} = A_{\text{Bone}} - \frac{\overline{CI} \cdot A_{\text{Muscle}}}{100}$$

where $\overline{\mathrm{CI}}$ is the mean CI for group 1 at the particular section level (Table 5). No significant correlation was found between RSD_{CI} and age, height, body weight, or L_{Tibia} for any section level, neither in group 1 nor in group 2 ($p > 0.1$).

Bending

The mean values of R_X and R_Y and their standard deviations are given in **Table 6**. As expected, the men in group 1 had higher values than the women. The female athletes of group 2, however,

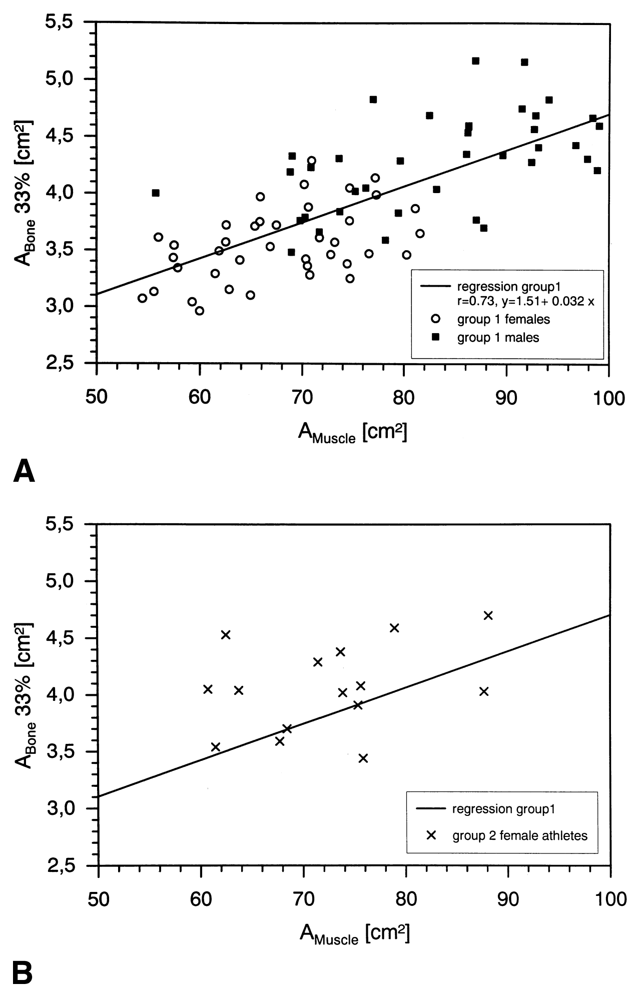


Figure 3. Regression of A_{Muscle} and A_{Bone14}. (a) Values obtained in healthy control subjects, aged between 20 and 30 years (group 1). The regression line is shown ($r = 0.73$). (b) Values obtained for female volleyball players (group 2). For comparison, the regression line of group 1 in (a) is depicted. The data points of group 2 are significantly above this line.

Table 4. Cross-sectional measures: Correlation coefficients

Group 1	A_{Muscle}	A_{Bone04}	A_{Bone14}	A_{Bone33}	A_{Bone66}
A_{Muscle}	—	0.66	0.74	0.73	0.70
A_{Bone04}	0.66	—	0.85	0.72	0.75
A_{Bone14}	0.74	0.85	—	0.91	0.90
A_{Bone33}	0.73	0.72	0.91	—	0.91
A_{Bone66}	0.70	0.75	0.90	0.91	—
A_{Bone66}	0.74	—	0.54	0.86	—

Correlation coefficients for measures of compressive force in men and women of group 1. All correlations were significant ($p < 0.001$).
KEY: *A*, area; subscript numbers indicate % level.

Table 5. Compression indices

	CI_{04} (%)	CI_{14} (%)	CI_{33} (%)	CI_{66} (%)
Group 1	5.25 ± 0.73	4.10 ± 0.46	5.25 ± 0.59	5.66 ± 0.67
Group 2	5.74 ± 0.84	4.34 ± 0.57	5.53 ± 0.70	6.33 ± 0.85

Mean ± SD of compression indices (CIs). Values for the different section levels are given in percent bone cross-sectional area per muscle cross-sectional area.

were indistinguishable from the men in group 1. Between section levels 14% and 66%, the bending strength increased, particularly around the *x* axis. Correlation analyses between *R* and MBM yielded the highest correlation coefficients at the 33% level. **Figure 5** depicts the data of R_X and MBM for the 33% section.

The different BIs of groups 1 and 2 are depicted in **Figure 6**. There was no difference in BI between women and men in group 1, nor between group 1 and the group 2 athletes, except for BI_{X66}, which was larger in group 2 than in the women of group 1. Because women and men in group 1 were not statistically different, the pooled means were computed and are given in **Table 7**. Over all subjects of group 1, there was no significant correlation of age, height, weight, or L_{Tibia} with any of the BIs, except for BI_{Y14}, which was negatively correlated with L_{Tibia} ($r = -0.34$, $p < 0.01$). The correlation R_{Y14} The correlation R_{Y14} with A_{Muscle} ($r = 0.71$) was still weaker than with MBM.

For group 1, the BI residuals were calculated in the same way as for the CI. Again, no significant correlation was found between residuals and age, height, or weight ($p > 0.1$). Only at the 14% level were the residuals significantly correlated with L_{Tibia} ($r = -0.34$ and $r = -0.40$ for *x* and *y* axes, respectively; $p < 0.01$).

Figure 7 depicts the material eccentricity, computed as R_{XBone}/A_{Bone} for the 33% and 66% section levels. Material eccentricity at 33% and 66% was correlated with L_{Tibia} ($r = 0.60$ and $r = 0.59$, respectively), but not at 14% ($p > 0.1$). Values for the 66% section were about twice those for the 33% section.

Table 8 gives the mean values of I_X and I_Y for groups 1 and 2. The values resemble those of R_X and R_Y, indicating that men had a greater moment of inertia than women of group 1, but not greater than the female athletes from group 2.

Table 6. Measures for bending

	14%	33%	66%
R_X (cm^3)			
Group 1 women	$0.88 \pm 0.12^{a,b}$	$1.07 \pm 0.17^{a,b}$	$1.77 \pm 0.25^{a,b}$
Group 1 men	1.16 ± 0.14	1.42 ± 0.21	2.27 ± 0.34
Group 2	1.09 ± 0.11	1.33 ± 0.24	2.42 ± 0.36
Correlation with MBM	0.77	0.81	0.74
R_Y (cm^3)			
Group 1 women	$0.88 \pm 0.12^{a,b}$	$0.86 \pm 0.13^{a,b}$	$1.27 \pm 0.18^{a,b}$
Group 1 men	1.18 ± 0.14	1.18 ± 0.16	1.69 ± 0.27
Group 2	1.08 ± 0.10	1.15 ± 0.13	1.67 ± 0.20
Correlation with MBM	0.77	0.78	0.68

Mean ± SD of the bone's area moment of resistance, *R*, around the *x* and *y* axes are shown. For all of group 1, correlation coefficients for the muscle bending moment (MBM) are given at the 14%, 33%, and 66% levels. Between 14% and 66%, bending strength increases by 98% around the *x* axis, but only by 44% around the *y* axis. Correlation with MBM was closest at the 33% level.
[a]Different from group 2.
[b]Different from group 1 men.

Discussion

The present data confirm hypothesis *H*1—that is, the existence of relevant relationships between muscle and bone. The simple observation that A_{Bone} as well as R_{Bone} were greater at higher section levels (i.e., with basically the same weight) again indicates that the influence of body weight is comparatively small. In our subjects, more than half of the variation in A_{Bone} could be attributed to the variation in A_{Muscle}. Moreover, roughly two thirds of the interindividual variation in R_{X33} could be explained by the muscle bending moment. Higher correlation coefficients for muscle-bone relations than in our study have been observed in children of different ages.[26] We ascribe this to two circumstances: (i) investigations in children consider a greater variance of the independent variable (i.e., A_{Muscle} is more widely spread); and (ii) children's muscles are still growing, which means that their bone is, in general, closer to the modeling than to the remodeling threshold.

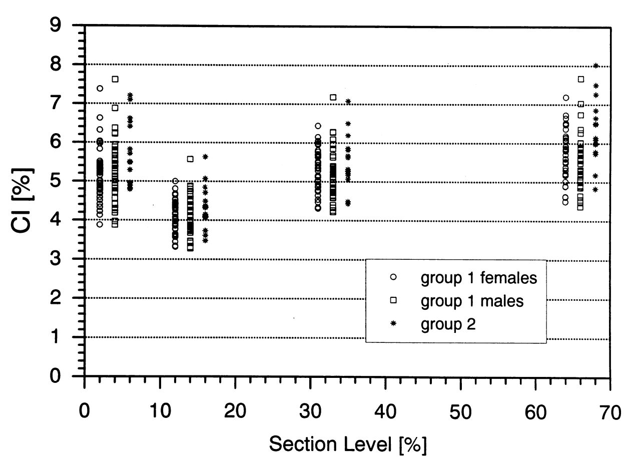


Figure 4. Compression indices of the lower leg. Compression indices (CIs) of the shank are depicted for the 4%, 14%, 33%, and 66% section levels. CI computed as $A_{Bone}/A_{Muscle} \cdot 100 \cdot A_{Muscle}$ is taken at 66%. At the 14% section level, CI was significantly lower than at all other levels ($p < 0.05$).

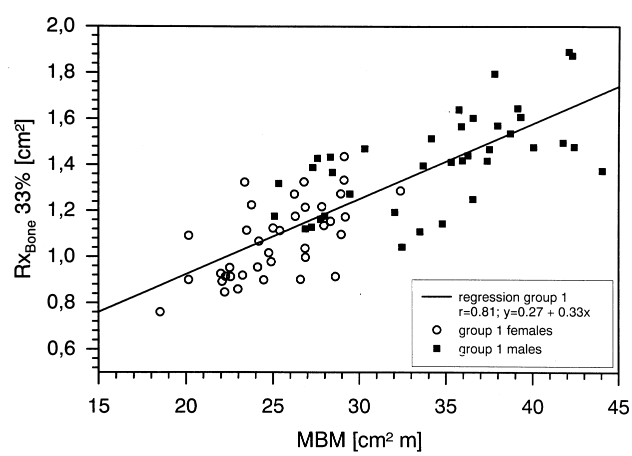
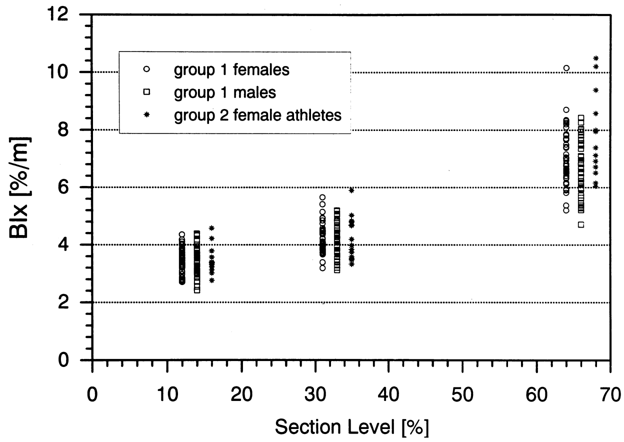


Figure 5. Bending moment and pQCT. pQCT data obtained from healthy control subjects (group 1). Regression analysis of muscle bending moment (MBM) and bone area moment of resistance to anteroposterior bending (R_X) at the 33% section level. MBM = $A_{Muscle} \cdot L_{Tibia} \cdot R_X$ is density-weighted. Correlation coefficient (r) = 0.81.



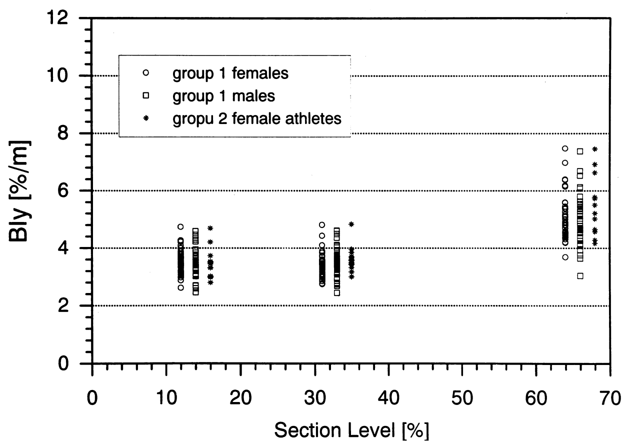


Figure 6. Bending indices. Bending indices (BIs) of the shank for R_{XBone} in (a), and R_{YBone} in (b). BI = R_{Bone}/MBM · 100. R_X stands for resistance to anteroposterior bending. Data include female and male young control subjects in group 1, and female athletes in group 2. While R_X and R_Y are almost equal at 14%, the 33% and 66% levels depict (i) increasing values, and (ii) an increasing R_X/R_Y ratio. Thus, the proximal shank seems to be particularly adapted to anteroposterior bending.

In this study, A_{Muscle} served as a surrogate for force production. We did not distinguish between foot extensors and flexors. Furthermore, it is known that a pennate muscle, like the soleus muscle, can produce a greater force per functional cross section than a fusiform muscle (e.g., the gastrocnemius). Different subjects may have had different proportions of their soleus and gastrocnemius CSA. There is general agreement, on the other hand, that different muscle fiber types, slow- or fast-twitch fibers, produce equal peak tension,[8] that the peak tension of muscles in men and in women is identical,[15,16,29] and that this is not influenced below the age of 60 years.[9,10] In contrast, athletes

Table 7. Bending indices

	14% level	33% level	66% level
BI_X (%/m)	3.46 ± 0.46	4.21 ± 0.54	6.89 ± 1.01
BI_Y (%/m)	3.52 ± 0.47	3.46 ± 0.48	5.05 ± 0.84

Mean ± SD of the bone's bending indices (BIs) around the *x* and *y* axes, as assessed from all data of group 1.

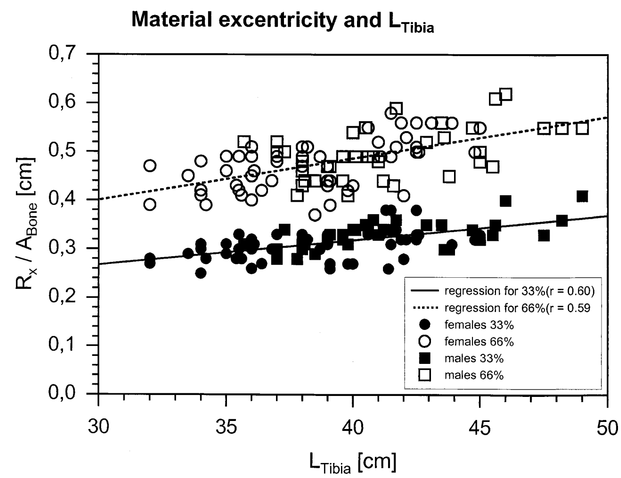


Figure 7. Material eccentricity and L_{Tibia}. Material eccentricity (ME) is computed as R_X/A_{Bone}. It provides a simple measure for the localization of bone mass away from its center. ME can serve as a measure for adaptation to bending in comparison to compression. The larger the ME, the better the bone adaptation to bending. ME was highest at 66%. A significant correlation between ME and L_{Tibia} was found at 33%, and 66%, but not at the 14% section level. Thus, a larger tibia is more than proportionally wider than a shorter one, most probably in order to make it more resistant to bending.

are believed to have a better recruitment of muscle force than ordinary subjects.[24] In conclusion, we would expect anatomical variation and the level of "fitness" to interfere with the peak force developed per unit of A_{Muscle}, but not gender, age, or fiber-type composition.

So far, only uniaxial compression has been considered. However, the human tibia is also adapted to bending, with the bending moment being caused by buckling (see Appendix) and by eccentric insertion of the muscles. At 66%, the knee extensors probably have a strong impact. This could explain why correlation coefficients for CI and BI are lower here than at the 33% section level. It would also explain the increased material eccentricity above the 50% section level (Figure 7); that is, the tibia beyond this level increases anteroposterior circumference by twofold, but its CSA remains almost constant. This cannot be understood as an adaptation to buckling, but rather to other bending moments (e.g., by eccentric muscle pull from the thigh).

In this study, the highest muscle/bone correlation as the R_X/MBM ratio was at 33%, where *r* was significantly greater than for A_{Bone}/A_{Muscle}. In other words, at the 33% level, bone architecture seems to be particularly adapted to anteroposterior bending moments.

We now examine the data of the professional volleyball players (*H*3). Their height and weight were indistinguishable from the men of group 1, but their shank muscles were significantly smaller. Consequently, they had a significantly elevated CI (see also Fig. 3B). We do not assume that their bones depict an elevated susceptibility to strain. Rather, the athletes seem to be able to recruit greater muscle tension than control subjects. Regarding bending measures, only the 66% level depicted a significantly greater BI in the athletes, whereas, at the 33% level, no significant difference was found ($p = 0.072$ and $p = 0.11$ for *x* and *y* axes, respectively).

Surprisingly, no gender-related difference could be found in the CI (hypothesis *H*2). This was true for the midshaft, epiphysis, and metaphysis. Our female and male subjects were all in good health. All women of group 1 had had a regular menses without

Table 8. Measures for buckling

	14% level		33% level		66% level	
	I_X (cm^4)	I_Y (cm^4)	I_X (cm^4)	I_Y (cm^4)	I_X (cm^4)	I_Y (cm^4)
Group 1						
Women	1.00 ± 0.20[a,b]	1.06 ± 0.22[a,b]	1.32 ± 0.31[a]	0.95 ± 0.19[a,b]	2.86 ± 0.61[a,b]	1.65 ± 0.37[a,b]
Men	1.41 ± 0.24	1.51 ± 0.27	1.88 ± 0.45	1.50 ± 0.29	4.02 ± 0.88	2.50 ± 0.58
Group 2	1.32 ± 0.23	1.43 ± 0.16	1.68 ± 0.41	1.25 ± 0.35	4.44 ± 0.90	2.30 ± 0.41

Density-weighted area moments of inertia.
[a]Different from group 2.
[b]Different from group 1 men.

oral contraceptives for at least 6 months prior to the study. The short-term error of the method (pQCT) was 1.15% for A_{Muscle} and 0.28% for A_{Bone}. The short-term error of the L_{Tibia} measurement was 3 mm. For the 66% level, we showed that a variation in the actual section level affects A_{Muscle} and A_{Bone} in the same direction. The short-term error of CI_{66} is therefore $<$1%. Based on whole-body DXA measurements, Ferreti et al. showed a $>$17% surplus of total body bone mass in postpubertal females. Such large differences were not seen in our data. It is therefore possible that gender differences of MBSI at the human lower leg are smaller or even nonexistent. One might argue that women produce lower muscle tension than men. Such a view contradicts the direct measurements of Vandervoort and McComas,[29] and it raises the question as to why DXA measurements show whole-body differences. Possibly, the so-called "metabolic bone" in women is stored in regions other than the leg (e.g., axial skeleton or distal arm).

In summary, we have shown that correlation of A_{Bone} with the compressive force generator, A_{Muscle}, was fairly close at the 14% and 33% levels. Correlation with the torque generator, MBM, was greatest at the 33% level. The reason for this is buckling, but probably also the insertion of the calf muscles in the proximal half.

Second, there was no relation between the CI's residuals and L_{Tibia}, suggesting that A_{Bone} was solely dependent on A_{Muscle}. In contrast, good correlations were found between the area moment of resistance, R, at the 33% and 66% section levels and MBM, the latter being a function of L_{Tibia}. This means that taller people have wider bones in their legs, with a thinner shell and a greater structural stiffness. Thus, it is evident that, besides force, length is an important factor in hollow bones and requires due consideration in patient assessment.

Third, different subjects can develop different peak muscle tensions according to their level of fitness and other factors, which likewise deserves consideration.

Acknowledgments: The authors are particularly grateful to Birthe Feilke and Hendrikje Börst from our center for clinical recruitment, and also to U. Brandis and J. Dames for their splendid assistance. Very special thanks to the Olympic Training Center in Berlin for cooperation in this and other studies. Last, but not least, we are deeply grateful to our subjects. Without their selfless contribution, this study would not have been possible.

Appendix

Modeling and Remodeling Thresholds

Given the correlation of A_{Bone} with A_{Muscle}, we computed CI as a muscle-bone strength index for compression. We now compare the results with the expected forces and thresholds for modeling and remodeling.[11,23,28] As for forces, Hettinger provided a survey of various studies, which estimate muscle peak tension at around 600 kPa (anatomical cross section). More specifically, in the calf, 650 kPa has been reported (p. 24 in Hettinger[15]).

Regarding modeling and remodeling thresholds, the following arguments apply. The section level with the least CSA was at 14%, amounting to 4.1% for group 1 (Table 5). Given a peak muscle tension of 650 kPa, and assuming that 60% of A_{Muscle} is effective during peak force production, the expected bone tension $P = 650 \cdot 60\%$ kPa/4.1% $\sim$ 9500 kPa. Given Young's modulus $E = 18 \cdot 109$ Pa,[17] the peak strain exerted by the muscle $\epsilon_{\mathrm{peak}} \sim$ 530 microStrain. The actual value may be slightly higher, because, in activities like the landing phase in jumping, the foot flexors also contract and also because an additional force of 25% is caused by the body weight. Nonetheless, the value obtained is within the range limited by the thresholds for modeling and remodeling, which are around 1000 μS and 100 μS, respectively, and seems to lie within the conservative range of the mechanostat.[14]

Buckling

Buckling is a deformation in which bending is caused by uniaxial load. In summer, for example, heavy ears of corn can make stalks buckle. According to Euler, buckling occurs if the compressive force:

$$F > \pi \cdot \frac{E \cdot I}{L^2}$$

where E is Young's modulus, I is the area moment of inertia, and L is the length of the beam. Let us consider I_{Bone} at 33%, which in group 1 men was 1.5 cm^4 for the y axis (Table 8). The mean L_{Tibia} was 41.3 cm (Table 1), and thus $L = 0.66 \cdot 41.3$ cm $= 27.3$ cm. Given $E = 18 \cdot 109$ Pa, as above, the buckling load is thus 1140 N. This load is easily reached by the A_{Muscle} of 83.1 cm^2 (Table 3). For the x axis (anteroposterior), the buckling load would be 1430 N.

The same calculation for the 14% section level yields a buckling load of 6500 N, which is probably beyond the maximum producible force. As a result, buckling, caused by muscle pull, definitely contributes to bending moments in the midshaft of the human tibia.

References

1. Biewener, A. A. Bone strength in small mammals and bipedal birds: Do safety factors change with body size? J Exp Biol 98:289–301; 1982.
2. Biewener, A. A. Biomechanics of mammalian terrestrial locomotion. Science 250:1097–1103; 1990.
3. Burr, D. B., Milgrom, C., Fyhrie, D., Forwood, M., Nyska, M., Finestone, A., Hoshaw, S., Saiag, E., and Simkin, A. In vivo measurement of human tibial strains during vigorous activity. Bone 18:405–410; 1996.

4. Demes, B., Stern, J. T. J., Hausman, M. R., Larson, S. G., McLeod, K. J., and Rubin, C. T. Patterns of strain in the macaque ulna during functional activity. Am J Phys Anthropol 106:87–100; 1998.
5. Ebbesen, E. N., Thomsen, J. S., and Mosekilde, L. Nondestructive determination of iliac crest cancellous bone strength by pQCT. Bone 21:535–540; 1997.
6. Ferretti, J. L., Capozza, R. F., Cointry, G. R., Garcia, S. L., Plotkin, H., Alvarez Filgueira, M. L., and Zanchetta, J. R. Gender-related differences in the relationship between densitometric values of whole-body bone mineral content and lean body mass in humans between 2 and 87 years of age. Bone 22:683–690; 1998.
7. Ferretti, J. L., Capozza, R. F., Cointry, G. R., Schneider, P., Reimers, C., and Schiessl, H. Bone mass is higher in women than in men per unit of muscle mass but bone mechanostat would compensate for the difference in the species. Bone 23(Suppl.):S471; 1998.
8. Fitts, R. H., McDonald, K. S., and Schluter, J. M. The determinants of skeletal muscle force and power: Their adaptability with changes in activity pattern. J Biomech 24(Suppl. 1):111–122; 1991.
9. Fitts, R. H., Troup, J. P., Witzmann, F. A., and Holloszy, J. O. The effect of ageing and exercise on skeletal muscle function. Mech Ageing Dev 27:161–172; 1984.
10. Frontera, W. R., Hughes, V. A., Lutz, K. J., and Evans, W. J. A cross-sectional study of muscle strength and mass in 45- to 78-yr-old men and women. J Appl Physiol 71:644–650; 1991.
11. Frost, H. M. Bone “mass” and the “mechanostat”: A proposal. Anat Rec 219:1–9; 1987.
12. Frost, H. M. Skeletal structural adaptations to mechanical usage (SATMU): 1. Redefining Wolff’s law: The bone modeling problem. Anat Rec 226:403–413; 1990.
13. Frost, H. M. Skeletal structural adaptations to mechanical usage (SATMU): 2. Redefining Wolff’s law: The remodeling problem. Anat Rec 226:414–422; 1990.
14. Frost, H. M. Wolff’s Law and bone’s structural adaptations to mechanical usage: An overview for clinicians. Angle Orthodont 64:175–188; 1994.
15. Hettinger, T. Isometrisches Muskeltraining. Stuttgart: Thieme; 1983.
16. Ikai, M. and Fukunaga, T. Calculation of muscle strength per unit cross-sectional area of human muscle by means of ultrasonic measurement. Int Z Angew Physiol 26:26; 1968.
17. Keller, T. S., Mao, Z., and Spenger, D. M. Young’s modulus, bending strength, and tissue physical properties of human compact bone. J Orthop Res 8:592–603; 1990.
18. Lanyon, L. E. Control of bone architecture by functional load bearing. J Bone Miner Res 7(Suppl. 2):S369–S375; 1992.
19. McArdle, W. D., Katch, F. I., and Katch, V. L. Body composition assessment. In: Balado, D., Eds. Exercise Physiology. Baltimore: Williams & Wilkins; 1991;541–575.
20. Nordstrom, P., Pettersson, U., and Lorentzon, R. Type of physical activity, muscle strength, and pubertal stage as determinants of bone mineral density and bone area in adolescent boys. J Bone Miner Res 13:1141–1148; 1998.
21. Powell, P. L., Roy, R. R., Kanim, P., Bello, M. A., and Edgerton, V. R. Predictability of skeletal muscle tension from architectural determinations in guinea pig hindlimbs. J Appl Physiol 57:1715–1721; 1984.
22. Russo, C. R., Taccetti, G., Caneva, P., Mannarino, A., Grimaldi, C., Maranghi, P., and Ricca, M. Musculoskeletal interactions in normal subjects and dialysis patients. Bone 23(Suppl.):S519; 1998.
23. Rubin, C. T. and Lanyon, L. E. Regulation of bone mass by mechanical strain magnitude. Calcif Tissue Int 37:411–417; 1985.
24. Sale, D. G. Neural adaptation to resistance training. Med Sci Sports Exerc 20(Suppl.):S135–S145; 1988.
25. Schiessl, H., Frost, H. M., and Jee, W. S. Estrogen and bone-muscle strength and mass relationships. Bone 22:1–6; 1998.
26. Schonau, E. The development of the skeletal system in children and the influence of muscular strength. Horm Res 49:27–31; 1998.
27. Schönau, E. and Frost, H. M. The muscle strength–bone strength relationship in humans: A review. In: International Congress on Osteoporosis, Xi’an, China; 1999; 84–89.
28. Turner, C. H., Forwood, M. R., Rho, J. Y., and Yoshikawa, T. Mechanical loading thresholds for lamellar and woven bone formation. J Bone Miner Res 9:87–97; 1994.
29. Vandervoort, A. A. and McComas, A. J. Contractile changes in opposing muscles of the human ankle joint with aging. J Appl Physiol 61:361–367; 1986.
30. Wilhelm, G., Felsenberg, D., Bogusch, G., Willnecker, J., Thaten, J., and Gummert, P. Biomechanical examinations for validation of the bone strength strain index SSI, calculated by peripheral quantitative computer tomography. In: Lyritis, G., Eds. Musculoskeletal Interactions. Athens: Hylonome; 1999; 105–108.
31. Wu, J., Ishizaki, S., Kato, Y., Kuroda, Y., and Fukashiro, S. The side-to-side differences of bone mass at proximal femur in female rhythmic sports gymnasts. J Bone Miner Res 13:900–906; 1998.
32. Yamada, H. Strength of Biological Materials. Baltimore: Williams & Wilkins; 1970.

Date Received: December 13, 1999
Date Revised: March 9, 2000
Date Accepted: April 17, 2000