ABSTRACT   **Objectives:** Morphological variation along the human limb reflects complex structural trade-offs between bone strength and mass. Here we assess how varying levels of plasticity and constraint affect this structure and influence the response to habitual loading along the diaphysis.

Materials and Methods: Cross-sectional geometric properties including total area, cortical area and rigidity were compared from the upper (humerus: 50% of length, radius: 66%, 50%, 4%) and lower (tibia: 50%, 38%, 4%) limbs of male varsity-level athletes and matched controls with distinct habitual loading histories.

Results: Geometric properties among cricketers and swimmers were significantly greater at the humeral midshaft, mid-proximal radius and radial midshaft compared to controls. By contrast, no significant differences were found among athletes or controls at the distal radius. The tibial midshafts of hockey players and runners also displayed greater area and rigidity compared to controls. Differences in geometry among the three groups became less pronounced distally, where structure was comparable among athletes and controls at 4% of tibial length. Additionally, coefficients of variation revealed that variation among athletes of the same sport was highest distally in both the upper and lower limb and lowest at midshaft, where structure most closely reflected the activity pattern of each loading group.

Discussion: These results support previous research suggesting that distal limb sections are more tightly constrained by safety factors compared to midshafts and proximal sections. Overall, it appears that plasticity and constraint vary not only between limb segments in correspondence to *known* activity patterns, but also along specific sections of the diaphysis.

Long bones are capable of adapting their shape and structure in response to mechanical stimulation through the apposition and resorption of skeletal tissue. In effort to quantify the relationship between habitual loading behavior and morphology, research focused on long bone adaptation has been conducted over the past several decades on bony vertebrates (Parfitt, 1979; Rubin and Lanyon, 1984; Frost, 1987; Mosley et al., 1997; Lieberman and Crompton, 1998; Hsieh et al., 2001; Judex et al., 2007; Demes and Carlson, 2009; Nicholson and Firth, 2010; Barak et al., 2011; Morimoto et al., 2011; Patel et al., 2013; Ruff et al., 2013). One aspect of study which calls for further inquiry is the formation and distribution of skeletal tissue along limb bone diaphyses that are exposed to markedly different forms of loading. While it is understood that for a given activity, levels of strain vary along long bone cortices (Lovejoy, 1976; Biewener, 1982; Biewener and Taylor, 1986; Drapeau and Streeter, 2001 Skerry, 2006), determining how bone sections respond to specific loading patterns requires further consideration. Comparing which regions of long bone diaphyses are more plastic to those that are more genetically canalized can therefore reveal how the limbs are influenced by biomechanical loading. If proximal and distal limb elements are indeed more susceptible to use and disuse related adaptation, varying degrees of plasticity and constraint should then be detectable along different diaphyseal sections as well.

Certain aspects of the appendicular skeleton including synovial joints and long bone epiphyses may be more adaptively constrained than diaphyses, which appear more phenotypically plastic by comparison (Ruff and Runestad, 1992; Ruff et al., 1993; Lovejoy et al., 1999, 2003; Lazenby et al., 2008). Previous research on biomechanical adaptation has also demonstrated that the middiaphyses of distal limb segments are typically the points of greatest peak bending during locomotion (Biewener, 1982; Biewener and Taylor, 1986). Accordingly,

studies centered on the relationship between mobility pattern and morphology often focus on the geometry of the midshaft (Trinkaus and Ruff, 1999a,b; Robling et al., 2006; Carlson et al., 2007; Wilks et al., 2009; Patel et al., 2013; Mongle et al., 2015a,b; but see Morimoto et al., 2011). Physiological loading is also capable of influencing specific regions of long bone diaphyses (Kannus et al., 1994; Adami et al., 1999), where aspects of the skeleton exposed to greater strain appear to elicit a stronger anabolic response for bone formation compared to locations farther from the primary source of strain (Heinonen et al., 2002). It has been established that the threshold for peak strain varies along weight-bearing bones as well, where a trigger for increased tissue formation closely corresponds with strain magnitude (Hsieh et al., 2001) as well as loading frequency and intensity (Fluck, 2006; Shaw and Stock, 2009b). Bone structure can also be influenced by a lack of mechanical stimulation, as histomorphometric analyses demonstrate that a greater degree of disuse related bone loss, following non-traumatic immobilization, is more prominent in the distal aspects of vertebrate limbs (Uhthoff et al., 1978, 1985). While prior studies have demonstrated that activity pattern is capable of influencing the cross-sectional geometry of discrete points along limbs (Rubin and Lanyon, 1984; Hsieh et al., 2001; Stock and Pfeiffer, 2001, 2004; Hallgrimsson et al., 2002; Marchi et al., 2006; Macintosh et al., 2014; Wallace et al., 2014), it is apparent that the limb as a functional unit is inclined to deposit greater mass proximally, while distal segments are more frequently remodeled and repaired to mitigate fatigue damage from the proportionately greater loads to which they are exposed (Hildebrand and Goslow, 2001; Pearson and Lieberman, 2004).

Skeletal adaptation and symmorphosis

Since Weibel and Taylor's observations of the economic trade-off between muscle density and oxygen consumption in the mammalian respiratory system (1981), the study of form-function relationships have expanded across biological systems and taxa to create the field of symmorphosis. Symmorphosis theory posits that the association between an organism's form - specifically the structure or size of a morphological character - should quantitatively correspond to the functional demands placed on it (Weibel et al., 1991). For many terrestrial vertebrates, limb morphology exhibits a tapering configuration from the proximal to distal segments so that greater mass (of bone and muscle tissue) is concentrated proximally, closer to the axis of rotation (Hildebrand and Hurley, 1985). Such an adaptation would have likely been selected for as a means of reducing energy expenditure while conserving angular momentum during locomotion, thus granting cursorial animals an advantage related to locomotor dexterity and bioenergetic efficiency (Alexander, 1981, 1998; Lieberman and Crompton, 1998; Lieberman and Pearson, 2001; Hildebrand and Goslow, 2001). Support for this premise has since been established experimentally in humans (Myers and Steudel, 1985; Dellanini et al., 2003). It has been demonstrated that applying greater mass along the distal lower limb requires an individual runner to increase oscillations during locomotion. In doing so, the runner must expend a greater amount of kinetic energy in order to reach the same speeds as another runner with greater mass located closer to their core (Dellanini et al., 2003).

While the tapered limb configuration may be beneficial energetically, it comes at an expense. Compared to more proximal sections, distal limb segments contain less cortical bone per unit space and are considerably more vulnerable to mechanical overload and fracture than their more proximal counterparts (Frost, 1997; Ruff, 2000; Rauch and Schoenau, 2001;

Plochocki et al., 2008; Shaw et al., 2014). To reduce the risk of structural failure, an optimal balance must be met between regional safety factors and function throughout the limb (Alexander, 1998; Skedros et al., 2003). Thoroughbred racehorses are a testament to how fine a trade-off exists between biomechanical strength and efficiency along regions of the fore- and hindlimb (Vaughan and Mason, 1975; Parkin et al., 2004; Verheyen et al., 2006). Nearly half of race-related forelimb injuries documented in a study by Vaughan and Mason (1975) were recorded at the metacarpus. Fractures declined proximally toward the humerus, where a mere 3.3% of injuries were reported among the horses. Similarly, 49.4% of female field hockey injuries were recorded between the ankle and tibia compared to 11.1% along the femur (Rose, 1981). A higher rate of strain-induced bone remodeling should be found distally on the limb, as relatively gracile skeletal elements closest to the point of surface impact lack the architectural strength of more proximal components (Currey, 1981; Burr et al., 1996). Indeed, among juvenile, sub-adult, and young adult sheep, remodeling rates were found to be higher at the midshaft of the distal limb (metatarsal and tibia) compared to the femoral midshaft (Lieberman et al., 2003). Just as bone distribution along distal limb segments appears to be optimized to endure typical loads, yet remain light enough to minimize energy expenditure, proximal elements are presumed to be less constrained by tissue economy (Stock and Pfeiffer, 2001, 2004). As such, proximal limb elements are often used to identify archaic and modern human behavioral patterns in the recent archaeological and fossil record (Trinkaus et al., 1994; Trinkaus et al., 1999; Marchi et al., 2006; Shackelford, 2007; Churchill and Rhodes, 2009; Havelkova et al., 2011; Trinkaus and Ruff, 2012). Femoral robusticity in particular has proven a useful indicator of mobility patterns in Holocene and Pleistocene hunter-gatherers (Ruff, 1999; Holt, 2003). The distal lower limb is also useful in reconstructing mobility in humans, however. Research conducted on the tibia

determined that the midshaft optimizes its structure under consistent mechanical loading (Trinkaus et al., 1994; MacDonald et al., 2009; Ireland et al., 2011; Shaw and Stock, 2013) while a study by Stock (2006) suggested that the tibial middiaphysis may better reflect human mobility than the femur.

A more comprehensive understanding of bone plasticity along the limb is necessary for an accurate interpretation of the fossil record, as more plastic segments should reflect adaptation to behavior while highly canalized regions may be more closely associated with taxonomic or genetic relationships (Waddington, 1942; Pigliucci, 2005; West-Eberhard, 2005; Young, 2006). Professional modern athletes are prime participants for study, as the subsistence behaviors employed by our ancestors such as long distance travel or overhand throwing (Bramble and Lieberman, 2004; Churchill and Rhodes, 2009; Roach et al., 2013), are reflected in modern sports. Moreover, athletic training generally begins during late childhood or adolescence, when new bone formation is highly responsive to load-induced strain (Kannus et al. 1995; Burr, 1997; Carter and Beaupre', 2001; Rauch and Schoenau, 2001; Raichlen et al., 2015). It is possible that subsistence practices among Upper and Lower Paleolithic hunter-gatherer populations would have begun at a relatively young age as well, as skillsets affiliated with foraging, and especially hunting, require both strength and experience (Hewlett and Cavalli-Sforza, 1986; Wiessner, 2002).

This study aims to advance the current understanding of site-specific long bone phenotypic plasticity by examining the relationship between diaphyseal structure and distinct forms of habitual loading in the upper and lower limbs. To address how skeletal plasticity and constraint vary along the limb, cross-sectional geometric properties indicative of mobility and behavior are quantified at various locations along the humeral, radial and tibial diaphyses of

male varsity-level athletes and matched controls. In doing so, plasticity and constraint are assessed along each diaphysis in relation to four habitual loading patterns.

MATERIAL AND METHODS

Subject sample

The cross-sectional images examined in this study were derived from a total of 83 males assigned to an upper ($n = 53$) and lower ($n = 51$) limb group. The humerus and radius were evaluated for the upper limb group which comprised swimmers (15), cricketers (17) and controls (21) while the lower limb group consisted of field hockey players (15), cross-country runners (15) and controls (21). Participants were between the ages of 19 and 30, all of which were recruited from the University of Cambridge and Anglia Ruskin University, Cambridge, U.K. The mean age of the recruited participants [hockey players (21.5, SD ± 1.5), runners (23.2, SD ± 3.2), swimmers (22, SD ± 2.5), cricketers (22.2, SD ± 2.5), and controls (21.5, SD ± 2.5)] was 22.1 years (SD ± 2.4). Competitive play and training amongst the athletes began during late childhood or early adolescence (mean age: hockey players: 9.6 years, runners: 13.7 years, swimmers: 10.5 years, cricketers: 9.5 years) by which all groups had since competed for 10.8 years on average (hockey players: 11.7 years, runners: 9.7 years, swimmers: 10.3 years, cricketers: 11.6 years). Of the 83 participants, 72 were right hand dominant. All subjects were confirmed non-smokers and had no prior history of limb fracture, disease or medication known to affect bone metabolism. Controls were healthy matched participants who led a relatively sedentary lifestyle (Booth, 2002), averaging less than one hour of strenuous physical activity per week. A questionnaire was issued to all subjects participating in the study as a means of obtaining athletic, lifestyle and medical history prior to any measurements being taken. Following completion of the

questionnaire, height was measured using a wall-mounted stadiometer (Seca, 770), while weight was measured with a digital scale (Seca, SEC888), for each participant. The study protocol was approved by the University of Cambridge Human Biology Research Ethics Committee and the Essex 2 Research Ethics Committee. Written informed consent was provided by each participant after receiving a verbal and written description of the protocol.

Cross-sectional imaging

For all participants of the lower limb group, three scans were taken along the left and right tibial diaphyses at 4%, 38%, and 50% of length, for a total of 306 pQCT images. Due to the asymmetric nature of bowling and throwing in the sport of cricket, elements of the right and left side of the upper limb were discriminated and scanned based on the lateral dominance of each individual in the upper limb group. A single scan was taken at 50% of total length of the dominant humerus (53 cross-sections) followed by three scans along the dominant radial diaphysis at 4%, 50%, and 66% of length (159 cross-sections), yielding a total of 212 upper limb cross-sectional images. The respective cross-section locations were selected to reflect the distal-most aspect, the forearm and plantar flexors and the midshaft of each element.

Two dimensional cross-sectional images were captured using non-invasive peripheral quantitative computed tomography (pQCT). The images were generated using an XCT 2000 (Stratec Medizintechnik GmbH, Pforzheim, Germany). Subjects assumed a comfortable sitting position facing the gantry opening while purpose-designed clamping devices were used to hold the limbs steady in a position parallel to the floor. Because femoral data was unavailable in the current study, plasticity and constraint were not evaluated along the entire lower limb as they were in the upper limb. Each scan provided a 2.5-mm-thick slice (0.5 pixel mm). The scan

locations of each long bone are illustrated in Figure 1. For additional details relating to data collection protocol, see Shaw and Stock (2009a,b).

[Figure 1 here]

The pQCT images were imported to ImageJ, a Java-based image processing program developed at the National Institutes of Health, Bethesda, Maryland, U.S (http://rsb.info.nih.gov/ij/). Each scan was analyzed using MomentMacro v.1.3; an ImageJ plug-in that calculates cross-sectional geometric properties (http://www.hopkinsmedicine.org/fae/mmacro.htm). Figures 2 and 3 provide examples of the upper and lower limb images examined along the three measured sections of the diaphyses.

Structural properties indicative of morphology, torsional rigidity and compressive/tensile strength were recorded along each segment of the diaphysis for the upper and lower limb to evaluate mechanical competence. The cross-sectional properties included cortical area (*CA*), an estimate of the bone's ability to buffer pure axial compression and tension; and total sub-periosteal area (*TA*), encompassing the periosteal contour as well as the medullary area of the section. The polar second moment of area (*J*), proportional to twice average bending rigidity in any two perpendicular planes (Ruff, 2008), was also calculated to infer rigidity along the diaphyses. To control for differences in body size and limb proportion across participants, surface areas (total area and cortical area) were standardized by body mass while the polar second moment of area was standardized by the product of body mass and moment arm length^2 (limb segment length) (Sparacello and Marchi, 2008).

[Figures 2, 3 here]

Additionally, relative values were calculated for each geometric property (*CA, TA, J*) by dividing cross-sectional property values from distal sections with matched property values from the proximal-most measured section of their respective limb [e.g., upper limb: (*TA* at 4% of radius / *TA* at 50% of humerus = relative *TA* at 4% of radius). These relative values provide an indication of diaphyseal structure at a location expected to display greater adaptive constraint relative to a location understood to exhibit relatively high plasticity (midshaft). In the upper limb, the humeral midshaft served as the proximal-most section while the tibial midshaft was used as the proximal-most section for comparison in the lower limb.

Analysis. Values for all geometric properties were calculated by averaging the right and left tibiae for all members of the lower limb group. Variation among the athlete and control groups was identified using univariate analysis of variance (ANOVA) where significance was recognized at $P \leq 0.05$ in all comparisons. After differences had been identified among the means, *post hoc* tests were used to classify the relationships of variance between groups. In instances that equal variance was assumed following Levene's test (homogeneity of variance), Hochberg's GT2 was consulted while the Games-Howell test was used where variance was unequal. The coefficient of variation (CV), defined as the ratio of the standard deviation to the mean (Hallgrimsson and Hall, 2005), was also calculated along segments of the diaphyses for *TA*, *CA*, and *J*. The CV was used for comparisons of inherent variation for area and rigidity among like-athletes within each loading group as a means of comparing intragroup variation

along each limb. To verify equality of variance, a parametric Levene's F test (P ≥ .05) was also conducted for CV along the diaphyses (Martin and Bridgmon, 2012).

RESULTS

Upper limb

ANOVA results, CVs and cross-sectional properties for the dominant upper limb of athletes and controls are reported in Tables 1-4. Geometric properties were compared among cricketers, swimmers and controls at the dominant humeral midshaft and at three locations along the dominant radius (4%, 50% and 66% of forearm length). Overall, the humeral midshaft of swimmers and cricketers was significantly stronger than those of controls. Swimmers displayed greater total sub-periosteal area (*TA*), cortical area (*CA*) and torsional rigidity (*J*) compared with controls at the humeral midshaft. Similarly, the dominant (throwing and bowling) arm of the cricketers displayed greater *TA*, *CA* and *J* at the humeral midshaft compared to controls (see Shaw and Stock, 2009a for detailed analysis on the non-dominant arm).

Comparisons along the forearm also yielded significant structural differences among the athletes and controls. The cricketers' dominant forearm displayed significantly greater *TA*, *CA*, and *J* at both the mid-proximal radius (66%) and midshaft compared to controls. By contrast, the swimmers' radial rigidity and cortical area were significantly greater than controls at midshaft but not at the mid-proximal radius. No differences in *TA*, *CA* or *J* were found at the distal radius (4%) in either cricketers or swimmers compared to controls.

[Table 1 here]

Upper limb; relative properties. At the dominant radial midshaft (50%), relative *TA* and *CA* were significantly lower in cricketers compared to controls, while at the mid-proximal radius (66%), only relative *CA* was significantly lower in cricketers. Comparisons involving the swimmer sample revealed that for both the 50% and 66% sites of the radii, relative *CA* was significantly lower compared to the control group, though no differences in rigidity or total rea were found. The smaller values associated with the relative properties along the radius emphasize the similarity in forearm structure between athletes and controls at the forearm compared to the stronger upper arms of the athletic sample.

Inherent variation along the upper limb. Coefficients of variation for *TA*, *CA* and *J* were calculated at the dominant humeral midshaft as well as the mid-proximal (66%), midshaft (50%) and distal (4%) radius using a combined sample that included cricketers, swimmers and controls. Considering the upper limb elements together, the CV for total area was lowest at the humeral midshaft (CV = 13.6) and highest at the distal radius (4%) (CV = 20.7) among athletes and controls. Along the forearm specifically, the radial midshaft displayed the least variation in area (CV = 15.5) compared to the distal or mid-proximal radius. Variation in rigidity and cortical area was also lowest at the humeral and radial midshafts in athletes and controls, while the greatest variation was found at the distal radius within each group.

Cricketers in particular displayed the greatest difference in variation for *TA* along the upper limb, where the CV was 11.9 at the radial midshaft and 27 at the distal radius. To place this in perspective, swimmers displayed a CV of 15.1 at midshaft and 20 at the distal radius while controls displayed a CV of 19.7 at midshaft and 15.3 at the distal radius. Comparisons

among groups revealed that the lowest level of inherent variation for rigidity (J) was found at the radial midshaft of controls (CV =23.2), while the highest level of variation was found at the distal radius of the cricketers' dominant (throwing) arm (CV = 53).

[Tables 2, 3 and 4 here]

Lower limb

Tables 5-7 summarize the one-way ANOVA comparisons, CVs and geometric properties measured along the tibiae of cross-country runners, field hockey players, and controls. Rigidity (J) at the tibial midshaft (50%) and mid-distal tibia (38%) was significantly higher in both runners and hockey players compared to controls, while no differences were found at the distal tibia (4%) among athletes or controls. Additionally, *CA* and *TA* were significantly higher in runners compared to controls at the tibial midshaft (50%) and mid-distal (38%) tibia. Comparisons of *TA* and *CA* at the mid-distal tibia and midshaft revealed no significant differences between hockey players and controls. Comparisons of *TA* and *CA* at the distal tibia (4%) revealed no significant differences among runners, hockey players or controls.

Lower limb; relative properties. Structural properties were compared along the tibial diaphysis to evaluate how distal segments vary in structure relative to the tibial midshaft. However, no significant differences were found for any relative properties at the mid-distal (38%) or distal (4%) tibia among the athletes and controls.

Inherent variation along the lower limb. Coefficients of variation (CV) for *TA*, *CA* and *J* were compared at midshaft (50%), mid-distal (38%) and distal (4%) sites along the tibial diaphysis using a combined sample that included distance runners, hockey players and controls. Inherent variation along the tibia was generally low (Simpson et al., 1960) compared to the upper limb, where CVs among groups ranged between 6.2 (*TA* 38% length) and 13.9 (*TA* 4% length). Overall, variation was lowest at midshaft and the mid-distal tibia while it was highest distally within the control and athletic groups. When compared separately, controls displayed the highest levels of inherent variation for *TA* (13.9), *CA* (19.6) and *J* (36) at the distal tibia compared to runners and hockey players, while the two athletic groups displayed the lowest levels of variation for *TA*, *CA* and *J* at midshaft and the mid-distal tibia.

[Tables 5, 6 and 7 here]

DISCUSSION

The primary aim of this study was to better understand how long bone cross-sectional morphology varies along the human upper and lower limb and whether this subsequently reflects phenotypic plasticity and constraint along the limbs. The results reveal that measures of bone strength and rigidity differ not only between limb elements but also at discrete points along the diaphyses. The humeral midshaft appears more disposed to adapt its structure in response to mechanical loading than any other site measured along the upper limb, suggesting that the upper

arm may be more plastic than the forearm. Analysis of the radius revealed that the morphology of the midshaft and mid-proximal diaphysis also reflected adaptation to habitual loading activities in cricketers and swimmers, while, by contrast, the distal radius displayed no structural differences among athletes or controls (see Fig. 4). The lack of morphological variation found at the distal radius compared to the radial midshaft, mid-proximal radius and humeral midshaft across groups may indicate that adaptive constraint increases with distance from midshaft. However, it remains unclear whether the relatively smaller muscle mass and limited loading exposure at distal compared to proximal limb segments influences structural variation along the diaphyses to a greater extent. Within the swimmer and cricketer samples, the lowest coefficients of variation for areas and rigidity were found at the humeral and radial midshafts, while inherent variation was highest at the distal radius. In other words, individuals within each athletic loading group were most morphologically similar to their fellow teammates at midshaft while displaying greater variation (CV) from one another distally. This observation suggests that midshafts may indeed be more inclined to adapt their structure compared to more proximal or distal sections when placed under a habitual loading regime. Moreover, sections a greater distance from midshaft may be more adaptively constrained.

Similar to the upper limb, it appears that rigidity and total cross-sectional area at the midshaft and mid-distal tibia better reflect morphological adaptation to running and hockey related loading patterns compared to the distal tibia. In fact, total area, cortical area and rigidity at the distal tibia were comparable among athletes and controls. The structural disparity between the midshaft and distal tibia may be indicative of a greater constraint of mass and tissue at more distal sections of the leg to prevent injury. Alternatively, this observation may reflect the imposition of maximal strains placed on the midshaft compared to more proximal or distal

sections a greater distance away. Intragroup coefficients of variation were also found to be greater distally in the lower limb sample while midshaft morphology was most similar among individuals within their respective groups. These results suggest that aspects of the upper limb may respond to mechanical stimulation in a similar manner to the lower limb despite their exposure to markedly different forms of loading, i.e., weight-bearing, ground impact forces in the lower limb vs. weight-supported, low resistance loading in the upper limb (Duncan et al., 2002).

[Figure 4 here]

Limb optimization

Vertebrate musculoskeletal systems operate close to their design limits, reflected by the high frequency of injury to organs closely associated with locomotion (i.e., torn ligaments, sprains, fractures) (Schmidt-Nielsen, 1984). To account for habitual loading behaviors that place varying degrees of strain along the limbs, a balanced trade-off between the strength of a given bone section and its mass should be met in the interest of energy conservation. Accordingly, safety factors (i.e., the ratio of maximum load capacity to typically endured loads) along the limb are not constant but vary by region (Alexander, 1981; 1998; Currey, 2002; Skerry, 2006). Though it is evident that distal elements do not model themselves to the same extent as more proximal elements, as cortical area would be comparable throughout the entirety of the limb (Lieberman, 2003), there remains some contention with respect to the rate that bone modeling and remodeling occur across individual limbs and elements. Human (Stock and Pfeiffer, 2001;

Weiss, 2003) and non-human primates (Walker, 1974; Bello-Hellegouarch et al., 2012) load their upper and lower limbs in a variety of ways during locomotion, which in turn, influences mobility and limb bone morphology differently than other mammals (Raichlen, 2006; Patel et al., 2013; Ruff et al., 2013). Some examples include increased measures of long bone length and robusticity, a relative lack of diaphyseal curvature compared to more mobile terrestrial mammals, longer strides in conjunction with a lower stride frequency, and larger distal limb segments necessary for manipulation of the hands and feet (Alexander and Maloiy, 1984; Swartz, 1990; Kimura, 1991; Polk et al., 1997; Jungers et al., 1998; Yamanaka et al., 2005; Drapeau and Streeter, 2006; Macintosh et al., 2015). Despite bearing relatively greater mass distally compared to cursorial specialists, both the upper and lower limbs of the athletic loading groups considered in this study displayed greater geometric differences from one another at midshaft and more proximal segments of their limbs than those distally. Because muscle force is understood to constitute the greatest mechanical load upon weight bearing bones (Frost and Schönau, 2000; Rittweger, 2008; but see Judex and Carlson, 2009), it is possible that the relatively small amount of muscle mass situated at the distal limb segments place a proportionately smaller strain along the diaphyses. It would follow that a smaller amount of microstrain acting on the distal limbs would fail to stimulate new bone formation (Turner and Pavalko, 1998; Frost, 2003), which may explain why so few geometric differences are observed in distal compared to proximal segments. The apparent canalization found at the 4% section in both the radius and tibia may relate more directly to the metaphyseal flaring found at these locations. While total area was nearly four times greater at the radial epiphysis than at midshaft in the athletic groups, cortical area was virtually the same between both locations. The presence of trabecular tissue at the distal and proximal sections of long bones may be responsible for

confining cortical area in spite of loading pattern, while the thicker cortical shell located at midshaft can be more readily distributed along the mediolateral (M-L) or anteroposterior (A-P) axes (where moments of bending and torsion are greatest). A side-by-side comparison emphasizing differences in total area and cortical area distribution at the tibial midshaft, mid-distal and distal tibiae of runners and hockey players can be viewed in Figure 5.

[Figure 5 here]

Just as muscle contractions constitute a mechanical load on bone (Özkaya et al., 2012), ground reaction force vectors are also capable of influencing skeletal morphology (Whalen, 1988; Kohrt et al., 1997; Crossley et al., 1999; Demes et al., 2001; Demes and Carlson, 2009; Wilks et al., 2009). Traditional cursorial mammals including red deer (Skedros et al., 2003), horses (Nicholson and Firth, 2010) and sheep (Lieberman, 2003) have been reported to exhibit more frequent episodes of haversian remodeling in the distal limb compared to more proximal segments. This phenomenon is likely due in part to the distal elements' exposure to shock, effectively buffering impact forces for the rest of the limb (Biewener, 1989; Skedros et al., 2003; Demes and Carlson, 2009; Wilks et al., 2009). The greater surface area of proximal and distal long bone sections may therefore be necessary to accommodate cancellous bone and cartilage for shock absorption without adding extraneous mass from cortical tissue. Unlike quadrupedal mammals, which rely on the forelimb for mobility, human upper limbs are not required to absorb high frequency strains generated during locomotion. In spite of ground reaction forces having limited influence on the human upper limb (Humphrey et al., 2008), the results presented here

reveal that geometry along the radius of swimmers and cricketers varied significantly to that of controls, except at the distal-most measured section of the radius (4% of length). Animals that do perform high frequency pronograde locomotion tend to exhibit increased remodeling episodes as a means of repairing microfractures in vulnerable distal elements, while stronger proximal elements appear to exhibit increased modeling of new bone (Lieberman and Crompton, 1998; Lieberman et al., 2003). Accordingly, the high level of structural variation in the humerus among athletes and controls corresponds with load-induced bone formation generated through modeling (Frost, 1997) while distal aspects of the upper limb likely undergo more infrequent modeling episodes. Though this may be the case, it is evident that modeling and remodeling are not restricted to limb elements but also vary by location along the diaphyses. In the lower limb, the distal tibiae of runners and field hockey players, which are regularly exposed to high intensity ground impact forces, also display little distinctive variation in rigidity and total area between groups. The observed lack of variation in the distal forearm may suggest that safety factors are still prioritized along the upper limb in humans as they are in cursors that do rely on the forelimb for locomotion. Moreover, the higher variation at midshaft and higher constraint distally along the limbs lends further support to the argument for canalization as a primary driver of distal forearm and shank structure in humans, while proximal segments of the upper and lower limb appear to adapt more readily in response to loading repertoire.

Regional geometry along the limbs

Dexterous use of the upper limb is essential in varsity sports such as cricket and swimming. The biomechanics that correspond with both sports separate them considerably, and

consequently, appear to influence upper limb morphology quite differently. Varsity swimmers engage in a number of swimming maneuvers including the front crawl, breast, freestyle, back and butterfly strokes; the bilateral upper limb oscillations of which are responsible for the majority of forward movement through the water (Seifert et al., 2004). Unlike swimming, most upper limb loading associated with cricket can be considered unilateral, where activities including batting, high velocity overhand throwing and bowling favor an individual's dominant arm (Elliot, 1993; Freeston et al., 2007).

The significantly greater rigidity of the cricketers' dominant humeri compared to controls in this study likely illustrates the impact of consistent bowling and batting loads placed on the upper arm from adolescence (Crockett et al., 2000; Levine et al., 2006; Neil and Schweitzer, 2008). The propensity of the humerus to plastically adapt its structure in response to high frequency loading is in agreement with the findings of Lieberman and Crompton (1998), where more proximal limb segments tend to be less constrained by tissue economy than their distal counterparts. Though the torque generated during throwing, bowling and batting play an apparent role in shaping humeral and radial midshaft morphology, the same activities appear to have a less obvious influence on the distal radius. Comparisons of cortical area and sub-periosteal area of the humerus relative to the radius of cricketers and swimmers further emphasize how the upper arm adapts to high frequency loading compared to distal segments of the limb, irrespective of loading pattern. Throwing related injuries like ulnar collateral ligament (UCL) tears often transpire due to disproportionate stress placed on the elbow and proximal forearm during high velocity throwing motions compared to more distal segments (Fleisig et al., 1995; Hariri and Safran, 2010; Podesta et al., 2013). The greater rigidity identified at the cricketer's humeral midshaft and mid-proximal radius may therefore result from the high strain

placed on the proximal portion of the upper limb during repetitive throwing, especially considering the swimmer's and controls' mid-proximal radial structure did not differ significantly by comparison. While loading pattern and frequency should not be overlooked as major drivers of skeletal adaptation (Judex et al., 2007; Ireland et al., 2011), it is apparent that tissue deposition at midshafts are more responsive to mechanical stimulation than proximal and distal sections of increasing distance. The smallest degree of inherent variation (CV) within each group was identified at midshaft as well, where the tibial midshafts of individual runners consistently conformed to a pattern of greater tissue distribution in the A-P plane. By contrast, hockey players generally allocated tissue across the M-L and A-P planes more evenly (Shaw and Stock, 2009a), though CVs were still lowest at midshaft compared to the distal tibiae. These structural patterns found at midshaft further indicate that the mid-tibia is morphologically specialized to a given loading pattern compared to the more uniform, distal aspects of the tibia exhibited in both athletes and controls.

Though field hockey and running are both associated with bilateral loading of the legs, the biomechanics involved in both sports are inherently different. The degree of lower limb mobility associated with ball games such as field hockey is intermittent and tends to fluctuate throughout the course of a match (Spencer et al., 2004; Macleod et al., 2009). Field hockey players attempt to preserve energy throughout the match by executing low intensity locomotor behavior ranging from light jogging to standing in position, only engaging in multi-directional, high intensity sprinting when necessary (Spencer et al., 2004). Successful marathon running is not so much dependent on power but endurance (Bramble and Lieberman, 2004; Palmer and Reid, 2009), and thus, an increase in loading frequency and repetition is substituted for a decrease in intensity. Unlike field hockey, cross-country running by and large assumes a

unidirectional mobility pattern, where the greatest strain is oriented in the A-P axis during forward locomotion (Macdonald et al., 2009; Marchi and Shaw, 2011). The aforementioned differences between the two lower limb activities may be responsible for influencing geometry along the distal tibial diaphysis as well, albeit to a lesser extent than at midshaft, though genetic canalization may ultimately be a greater regulator of metaphyseal and epiphyseal morphology (Lovejoy et al., 1999, 2003; Lazenby et al., 2008). Bearing these results in mind, interpreting behavior along both the upper and lower limbs may be most effective by studying cross-sectional properties at or near the midshaft, as activity pattern among individuals and populations becomes more difficult to discern with increasing distance from these regions.

ACKNOWLEDGMENTS

LITERATURE CITED

Adami S, Gatti D, Braga V, Bianchini D, Rossini, M. 1999. Site-specific effects of strength training on bone structure and geometry of ultradistal radius in postmenopausal women. J Bone Miner Res 14:120–124.

Alexander RM. 1981. Factors of safety in the structure of mammals. Sci Prog 67:109–130.

Alexander RM. 1998. Symmorphosis and safety factors. In: Weibel E, Taylor C, Bolis L, editors. Principles of animal design: the optimization and symmorphosis debate. Cambridge: Cambridge University Press. p 28-36.

Alexander RMN, Maloiy GMO. 1984. Stride lengths and stride frequencies of primates. J Zoo Lond 202:577–582.

Barak MM, Lieberman DE, Hublin JJ. 2011. A Wolff in sheep's clothing: Trabecular bone adaptation in response to changes in joint loading orientation. Bone 49:1141–1151.

Bello-Hellegouarch G, Potau JM, Arias-Martorell J, Pastor JF, Diogo R, Perez-Perez A. 2012. The Rotator Cuff Muscles in Hominoidea: Evolution and Adaptations to Different Types of Locomotion. In: Hughes EF, Madison HE, editors. Primates: Classification, Evolution and Behavior. New York: Nova Science Publishers. p 111-134.

Bertram JEA, Biewener AA. 1988. Bone Curvature: Sacrificing Strength For Load Predictability? J Theor Biol 131:75–92.

Biewener AA. 1982. Bone Strength in Small Mammals and Bipedal Birds : Do Safety Factors Change With Body Size? J Exp Biol 98:289–301.

Biewener AA, Taylor CR. 1986. Bone strain: a determinant of gait and speed? J Ep Biol 123:383–400.

Biewener AA. 1989. Scaling body support in mammals: limb posture and muscle mechanics. Science 245:45-48.

Booth F. 2002. Cost and consequences of sedentary living: new battleground for an old enemy. President's Council Phys Fitness Sports Res Digest 3:1–8.

Bramble DM, Lieberman DE. 2004. Endurance running and the evolution of Homo. Nature 432:345–52.

Burr DB, Milgrom C, Fyhrie D, Forwood M, Nyska M, Finestone A, Hoshaw S, Saiag E, Simkin A. 1996. In vivo measurement of human tibial strains during vigorous activity. Bone 18:405–410.

Burr DB. 1997. Muscle Strength, Bone Mass, and Age-Related Bone Loss. J Bone Miner Res 12:1547–1551.

Carlson KJ, Grine FE, Pearson OM. 2007. Robusticity and Sexual Dimorphism in the Postcranium of Modern Hunter-Gatherers From Australia. Am J Phys Anthropol 23:9–23.

Churchill SE, Rhodes JA. 2009. The Evolution of the Human Capacity for "Killing at a Distance": The Human Fossil Evidence for the Evolution of Projectile Weaponry. In: Hublin JJ, Richards MP, editors. The Evolution of Hominin Diets: Integrating Approaches to the Study of Palaeolithic Subsistence. Houten: Springer Netherlands. p 201–210.

Crockett HC, Gross LB, Wilk KE, Schwartz ML, Reed J, O'Mara J, Reilly MT, Dugas JR, Meister K, Lyman S, Andrew JR. 2002. Osseous Adaptation and Range of Motion at the Glenohumeral Joint in Professional Baseball Pitchers. Am J Sports Med. 30:20-26.

Currey JD. 1981. What is bone for? Property-function relationships in bone. In: Cowin SC, editor. Mechanical properties of bone. New York: American Society of Mechanical Engineers. p 13–26.

Currey JD. 2002. Bones: structure and mechanics. Princeton: Princeton University Press.

Dellanini L, Hawkins D, Martin RB, Stover S. 2003. An investigation of the interactions between lower-limb bone morphology, limb inertial properties and limb dynamics. J Biomech 36:913–919.

Demes B, Qin YX, Stern JT, Larson SG, Rubin CT. (2001). Patterns of strain in the macaque tibia during functional activity. Am J Phys Anthropol, 116:257–65.

Demes B, Carlson KJ. 2009. Locomotor variation and bending regimes of capuchin limb bones. Am J Phys Anthropol. 139:558–71.

Drapeau MSM, Streeter MA. 2006. Modeling and remodeling responses to normal loading in the human lower limb. Am J Phys Anthropol. 129:403–9.

Duncan CS, Blimkie CJR, Kemp A, HiggsW, Cowell CT,Woodhead H, Briody JN, Howman-Giles R. 2002. Mid-femur geometry and biomechanical properties in 15- to 18-yr-old female athletes. Med Sci Sports Exerc 34:673–681.

Elliot B, Baker J, Foster D. 1993. The kinematics and kinetics of the off-drive and the on-drive in cricket. Am J Biomed Sci 25:48-54.

Fleisig GS, Andrews JR, Dillman CJ, Escamilla RF. 1995. Kinetics of baseball pitching with implications about injury mechanisms. Am J Sports Med 23:233–239.

Flück M. 2006. Functional, structural and molecular plasticity of mammalian skeletal muscle in response to exercise stimuli. J Exp Biol 209:2239–48.

Freeston J, Ferdinands R, Rooney K. 2007. Throwing velocity and accuracy in elite and subelite cricketer players: A descriptive study. Eur J Sports Sci 7:231-237.

Frost H. 1987. The Mechanostat: A proposed pathogenetic mechanism of osteoporoses and the bone mass effects of mechanical and nonmechanical agents. J Bone Miner Res 2:73–85.

Frost HM. 1997. Why Do Marathon Runners Have Less Bone Than Weight Lifters? A Vital-Biomechanical View and Explanation. Bone 20:183–189.

Frost HM. 2003. Bone ' s Mechanostat : A 2003 Update. Anat Rec A Discov Mol Cell Evol Biol 275:1081–1101.

Frost HM, Schönau E. 2000. The "Muscle-Bone Unit" in Children and Adolescents: A 2000 Overview. J Pediatr Endocrinol Metab.

Hallgrimsson B, Willmore K, Hall BK. 2002. Canalization, developmental stability, and morphological integration in primate limbs. Am J Phys Anthropol 119:131–158.

Hallgrimsson B, Hall BK. 2005. Variation and variability: Central concepts in biology. Burlington, MA: Elsevier Academic Press.

Hariri S, Safran MR. 2010. Ulnar collateral ligament injury in the overhead athlete. Clin Sports Med 29:619-644.

Havelková P, Villotte S, Velemínský P, Poláček L, Dobisíková M. 2011. Enthesopathies and activity patterns in the Early Medieval Great Moravian population: Evidence of division of labour. Int J Osteoarchaeol 21:487–504.

Heinonen A, Sievänen H, Kannus P, Oja P, Vuori I. 2002. Site-specific skeletal response to long-term weight training seems to be attributable to principal loading modality: a pQCT study of female weightlifters. Calcif Tissue Int. 70:469–474.

Hewlett B, Cavalli-Sforza, LLL. 1986. Cultural transmission among Aka pygmies. Am Anthropol 88:922–934.

Hildebrand M. 1985. Walking and running. In: Hildebrand M, Bramble DM, Liem KF, Wake DB, editors Functional vertebrate morphology Cambridge, MA: Belknap Press. p 38–57.

Hildebrand M, Goslow GE Jr. 2001. Analysis of vertebrate structure. New York: John Wiley & Sons, Inc.

Holt BM. 2003. Mobility in upper paleolithic and mesolithic Europe: evidence from the lower limb. Am J Phys Anthropol 122:200–215.

Hsieh Y, Robling AG., Ambrosius WT, Burr DB, Turner CH. 2001. Mechanical Loading of Diaphyseal Bone In Vivo: The Strain Threshold for an Osteogenic Response Varies with Location. J Bone Miner Res 16:2291–2297.

Humphrey N, Mundlos S, Türkmen S. 2008. Genes and quadrupedal locomotion in humans. PNAS 105(21).

Ireland A, Korhonen M, Heinonen A, Suominen H, Baur C, Stevens S, Rittweger J. 2011. Side-to-side differences in bone strength in master jumpers and sprinters. J Musculoskelet Neuronal Interact 11:298–305.

Judex S, Carlson KJ. 2009. Is Bone's Response to Mechanical Signals Dominated by Gravitational Loading? Med Sci Sports Exerc 41:2037–2043.

Judex S, Lei X, Han D, Rubin C. 2007. Low-magnitude mechanical signals that stimulate bone formation in the ovariectomized rat are dependent on the applied frequency but not on the strain magnitude. J Biomech 40:1333–9.

Jungers WL, Burr DB, Cole M. 1998. Body Size and Scaling of Long Bone Geometry, Bone Strength, And Positional Behaviour in Cercopithecoid Primates. In: Strasser E, editor. Primate Locomotion. New York: Springer Science and Business Media. p 309–330.

Kannus P, Haapasalo H, Sieva̤nen H, Oja P, Vuori I. 1994. The site-specific effects of long-term unilateral activity on bone mineral density and content. Bone 15:279–284.

Kannus P, Haapasalo H, Sankelo M, Sieva̤nen H, Pasanen M, Heinonen A, Oja P, Vuori I. 1995. Effect of starting-age of physical activity on bone mass in the dominant arm of tennis and squash players. Ann Intern Med 123:27–31.

Kimura T. 1991. Long and robust bones of primates. In: Ehara A, Kimura T, Takenaka O, Iwamoto M, editors Primatology Toda. New York: Elsevier. p 495–498.

Kohrt WM, Ehsani AA, Birge, SJ. 1997. Effects of Exercise Involving Predominantly Either Joint-Reaction or Ground-Reaction Forces on Bone Mineral Density in Older Women. J Bone Miner Res 12:1253-1261.

Lazenby RA, Cooper DML, Angus S, Hallgrímsson B. 2008. Articular constraint, handedness, and directional asymmetry in the human second metacarpal. J Hum Evol 54:875–885.

Levine WN, Brandon ML, Stein BS, Gardner TR, Bigliani LU, Ahmad CS. 2006. Shoulder adaptive changes in youth baseball players. J Shoulder Elb Surg 15:562-566.

Lieberman, DE. 2003. Optimization of bone growth and remodeling in response to loading in tapered mammalian limbs. J Exp Biol 206:3125–3138.

Lieberman DE, Crompton AW. 1998. Responses of bone to stress: constraints on symmorphosis. In: Webel ER, Taylor CR, Bolis L, editors. Principles of animal design: the optimization and symmorphosis debate. Cambridge: Cambridge University Press. p 78–86.

Lieberman DE, Pearson OM. 2001. Trade-off between modeling and remodeling responses to loading in the mammalian limb. Bull Mus Comp Zool 156:269–282.

Lovejoy CO, Burstein AH, Heiple KG. 1976. The biomechanical analysis of bone strength: a method and its application to platycnemia. Am J Phys Anthropol 44:489–505.

Lovejoy CO, Cohn MJ, White TD. 1999. Morphological analysis of the mammalian postcranium: a developmental perspective. Proc Natl Acad Sci USA 96:13247–13252.

Lovejoy CO, McCollum MA, Reno PL, Rosenman BA. 2003. Developmental biology and human evolution. Annu Rev Anthropol 32:85–109.

Macdonald HM, Cooper DML, McKay HA. 2009. Anterior-posterior bending strength at the tibial shaft increases with physical activity in boys: evidence for non-uniform geometric adaptation. Osteoporos Int 20:61–70.

Macintosh AA, Pinhasi R, Stock JT. 2014. Lower limb skeletal biomechanics track long-term decline in mobility across ~6150 years of agriculture in central Europe. J Archaeo Sci 52:376-390.

Macintosh AA, Davies TG, Pinhasi R, Stock JT. 2015. Declining tibial curvature parallels ~6150 years of decreasing mobility in central european agriculturalists. Am J Phys Anthropol 157:260-275.

Marchi D, Sparacello V, Holt BM, Formicola V. 2006. Biomechanical approach to the reconstruction of activity patterns in Neolithic Western Liguria, Italy. Am J Phys Anthropol 131: 447–455.

Marchi D, Shaw CN. 2011. Variation in fibular robusticity reflects variation in mobility patterns. J Hum Evol 61:609–16.

Martin WE and Bridgmon KD. 2012. Quantitative and statistical research methods: from hypothesis to results Vol. 42. San Frtancisco: John Wiley & Sons. p 1-14.

Mongle CS, Wallace IJ, & Grine FE. 2015. Cross-sectional structural variation relative to midshaft along hominine diaphyses. I. The forelimb. Am J Phys Anthropol, in press.

Mongle CS, Wallace IJ, & Grine FE. 2015. Cross-sectional structural variation relative to midshaft along hominine diaphyses. II. The hindlimb. Am J Phys Anthropol, in press.

Morimoto N, De León MSP, Zollikofer CPE. 2011. Exploring Femoral Diaphyseal Shape Variation in Wild and Captive Chimpanzees by Means of Morphometric Mapping: A Test of Wolff's Law. Anat Rec 294:589–609.

Mosley JR, March BM, Lynch J, Lanyon LE. 1997. Strain magnitude related changes in whole bone architecture in growing rats. Bone 20:191–198.

Myers MJ, Steudel K. 1985. Effect of limb mass and its distribution on the energetic cost of running. J Exp Biol 116:363–73.

Neil JM, Schweitzer ME. 2008. Humeral cortical and trabecular changes in the throwing athlete: a quantitative computed tomography study of male college baseball players. J Comput Assist Tomogr 32:492-496.

Nicholson CL, Firth EC. 2010. Assessment of bone response to conditioning exercise in the radius and tibia of young thoroughbred horses using pQCT. J Musculoskelet Neuronal Interact 10:199–206.

Özkaya NN, Nordin M, Goldsheyder D, Leger D. 2012. Fundamentals of biomechanics: equilibrium, motion and deformation. New York: Springer Science & Business Media. p 165-185.

Palmer G, Reid A. 2009. Running: From Middle Distance to Marathon. London: A&C Black Publishers Ltd.

Parfitt AM. 1979. Quantum concept of bone remodeling and turnover: implications for the pathogenesis of osteoporosis. Calcif Tissue Int 28:1–5.

Parkin TDH, Clegg PD, French NP, Proudman CJ, Riggs CM, Singer ER, Morgan KL. 2004. Race-and course-level risk factors for fatal distal limb fracture in racing Thoroughbreds. Equine Vet J 36:521-526.

Patel BA, Ruff CB, Simons ELR, Organ JM. 2013. Humeral cross-sectional shape in suspensory primates and sloths. Anat Rec 296:545–56.

Pearson OM, Lieberman DE. 2004. The aging of Wolff's "law": ontogeny and responses to mechanical loading in cortical bone. Am J Phys Anthropol 39:63–99.

Pigliucci M. 2005. Evolution of phenotypic plasticity: where are we going now? Trends Ecol Evol 20:481–6.

Plochocki JH, Rivera JP, Zhang C, Ebba SA. 2008. Bone Modeling Response to Voluntary Exercise in the Hindlimb of Mice. J Morphol 318:313–318.

Podesta L, Crow SA, Volkmer D, Bert T, Yocum LA. 2013. Treatment of Partial Ulnar Collateral Ligament Tears in the Elbow With Platelet-Rich Plasma. Am J Sports Med 41:1689-1694.

Polk JD, Demes B, Jungers WL, Heinrich RE, Biknevicius AR, and Runestad JA. 1997. Cross-sectional properties of primate and nonprimate limb bones. Am J Phys Anthropol Suppl 24:188.

Raichlen DA, Gordon AD, Foster AD, Webber JT, Sukhdeo SM, Scott RS, Ryan TM. 2015. An ontogenetic framework linking locomotion and trabecular bone architecture with applications for reconstructing hominin life history. J Hum Evol 81:1–12.

Rauch F, Schoenau E. 2001. The developing bone: slave or master of its cells and molecules? Ped Res 50:309–14.

Rittweger J. 2008. Ten years muscle-bone hypothesis: What have we learned so far ? -Almost a Festschrift: A historical perspective. J Musculoskelet Neuronal Interact 8:174–178.

Roach NT, Venkadesan M, Rainbow MJ, Lieberman DE. 2013. Elastic energy storage in the shoulder and the evolution of high-speed throwing in Homo. Nature 498:483–486.

Robling AG, Castillo AB, Turner CH. 2006. Biomechanical and Molecular Regulation of Bone Remodeling. Annu Rev Biomed Eng. 8:455-495.

Rose CP. 1981. Injuries in women's field hockey: a four year study. Phys Sports Med 9:97-98, 100.

Rubin CT, Lanyon LE. 1984. Regulation of bone formation by applied dynamic loads. J Bone Joint Surg 66:397–402.

Ruff CB. 2000. Body size, body shape, and long bone strength in modern humans. J Hum Evol 38:269–290.

Ruff CB. 2008. Biomechanical Analyses Of Archaeological Human Skeletons. In: Katzenberg AM, Saunders SR, editors Biological Anthropology of the Human Skeleton, 2nd Ed. Hoboken, NJ: Wiley. p 183–206.

Ruff CB, Runestad JA. 1992. Primate limb bone structural adaptations. Annu Rev Anthropol 21:407–433.

Ruff CB, Trinkaus E, Walker A, Larsen CS. 1993 Postcranial robusticity in Homo. I: temporal trends and mechanical interpretation. Am J Phys Anthropol 91:21–53.

Ruff CB, Burgess ML, Bromage TG, Mudakikwa A, McFarlin SC. 2013. Ontogenetic changes in limb bone structural proportions in mountain gorillas (Gorilla beringei beringei). J Hum Evol 65:693–703.

Schmidt-Nielsen K. 1995. Scaling: why is animal size so important? Cambridge: Cambridge University Press. p 42-56.

Seifert L, Chollet D, Bardy BG. 2004. Effect of swimming velocity on arm coordination in the front crawl: a dynamic analysis. J Sports Sci 22:651-660.

Shackelford LL. 2007. Regional Variation in the Postcranial Robusticity of Late Upper Paleolithic Humans. Am J Phys Anthropol. 133:655–668.

Shaw CN, Stock JT. 2009a. Habitual throwing and swimming correspond with upper limb diaphyseal strength and shape in modern human athletes. Am J Phys Anthropol 140:160–172.

Shaw CN, Stock JT. 2009b. Intensity, repetitiveness, and directionality of habitual adolescent mobility patterns influence the tibial diaphysis morphology of athletes. Am J Phys Anthropol 140:149–59.

Shaw CN, Stock JT, Davies TG, Ryan TM. 2014. Does the Distribution and Variation in Cortical Bone Along Lower Limb Diaphyses Reflect Selection for Locomotor Economy?. In: Carlson KJ, Damainao M, editors Reconstructing Mobility. New York: Springer US. p 49-66.

Simpson GG, Roe A, Lewontin RC. 1960. Quantitative zoology. New York: Harcourt, Brace & Co.

Skedros JG, Sybrowsky CL, Parry TR, Bloebaum RD. 2003. Regional differences in cortical bone organization and microdamage prevalence in Rocky Mountain mule deer. Anat Rec A Discov Mol Cell Evol Biol 274:837–50.

Skerry TM. 2006. One mechanostat or many? Modifications of the site-specific response of bone to mechanical loading by nature and nurture. J Musculoskelet Neuronal Interact 6:122–7.

Sparacello V, Marchi D. 2008. Mobility and subsistence economy: a diachronic comparison between two groups settled in the same geographic area (Liguria, Italy). Am J Phys Anthropol 136:485–495.

Spencer M, Lawrence S, Rechichi C, Bishop D, Goodman C. 2004. Time-motion analysis of elite field hockey, with special reference to repeated-sprint activity. J Sports Sci. 22:834-850.

Stock JT. 2006. Hunter-gatherer postcranial robusticity relative to patterns of mobility, climatic adaptation, and selection for tissue economy. Am J Phys Anthropol 131:194–204.

Stock J, Pfeiffer S. 2001. Linking structural variability in long bone diaphyses to habitual behaviors: foragers from the southern African Later Stone Age and the Andaman Islands. Am J Phys Anthropol 115:337–48.

Stock JT, Pfeiffer SK. 2004. Long bone robusticity and subsistence behaviour among Later Stone Age foragers of the forest and fynbos biomes of South Africa. J Archaeol Sci 31:999–1013.

Swartz SM. 1990. Curvature of the forelimb bones of anthropoid primates: overall allometric patterns and specializations in suspensory species. Am J Phys Anthropol 83:477–498.

Taylor CR, Weibel ER. 1981. Design of the mammalian respiratory system. I. Problem and strategy. Respir Physiol. 44:1–10.

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.

Trinkaus E, Ruff CB. 1999a. Diaphyseal cross-sectional geometry of Near Eastern Middle Paleolithic humans: The femur. J Archaeol Sci 26:409–424.

Trinkaus E, Ruff CB. 1999b. Diaphyseal cross-sectional geometry of Near Eastern Middle Paleolithic humans: The tibia. J Archaeol Sci 26:1289–1300.

Trinkaus E, Ruff CB. 2012. Femoral and Tibial Diaphyseal Cross-Sectional Geometry in Pleistocene Homo. PaleoAnthropology 2012 13–62.

Trinkaus E, Stringer CB, Ruff CB, Hennessy RJ, Roberts MB, Parfitt SA. 1999. Diaphyseal cross-sectional geometry of the Boxgrove 1 Middle Pleistocene human tibia. J Hum Evol 37:1–25.

Turner CH, Pavalko FM. 1998. Mechanotransduction and functional response of the skeleton to physical stress: The mechanisms and mechanics of bone adaptation. J Orthop Sci 3:346–355.

Uhthoff HK, Jaworski ZF. 1978. Bone loss in response to long-term immobilisation. J Bone Joint Surg Br 60:420-429.

Uhthoff HK, Sékaly G, Jaworski ZF. 1985. Effect of long-term nontraumatic immobilization on metaphyseal spongiosa in young adult and old beagle dogs. Clin Orthop Relat Res. 192:278-283.

Vaughan LC, Mason BJE. 1975. A clinico-pathological study of racing accidents in horses. Dorking: Bartholomew Press.

Verheyen K, Price J, Lanyon L, Wood J. 2006. Exercise distance and speed affect the risk of fracture in racehorses. Bone 39:1322-1330.

Walker AC. 1974. Locomotor adaptations in past and present prosimian primates. In: Jenkins Jr. FA, editor Primate locomotion. New York: Academic Press. p 349-381.

Wallace IJ, Demeds B, Mongle C, Pearson OM, Polk JD, Lieberman DE. 2014. Exercise-induced bone formation is poorly linked to local strain magnitude in the sheep tibia. PloS One 9:e99108.

Weibel ER, Taylor CR, Hoppeler H. 1991. The concept of symmorphosis: a testable hypothesis of structure-function relationship. PNAS 88:10357–10361.

Weiss E. 2003. Effects of rowing on humeral strength. Am J Phys Anthropol 121:293–302.

West-Eberhard MJ. 2005. Phenotypic Accommodation: Adaptive Innovation Due to Developmental Plasticity. J Exp Zool 304B:610–618.

Whalen RT, Carter DR, Steele CR. 1988 Influence of physical activity of the regulation of bone density. J Biomech 21:825-838.

Wiessner P. 2002. Hunting, healing, and hxaro exchange. A long-term perspective on !Kung (Ju/'hoansi) large-game hunting. Evol Hum Behav 23:407–436.

Wilks DC, Winwood K, Gilliver SF, Kwiet A, Chatfield M, Michaelis I, Rittweger J. 2009. Bone mass and geometry of the tibia and the radius of master sprinters, middle and long distance runners, race-walkers and sedentary control participants: a pQCT study. Bone 45:91–7.

Yamanaka A, Gunji H, Ishida H. 2005. Curvature, length, and cross-sectional geometry of the femur and humerus in anthropoid primates. Am J Phys Anthropol 127:46–57.

Young NM. 2006. Function, ontogeny and canalization of shape variance in the primate scapula. J Anat 209:623–36.

FIGURE LEGEND

Fig. 1. The long bones analyzed in this study and their corresponding pQCT scan locations.

From left to right: humerus, 50% of length: radius, 66%, 50% and 4% of length: tibia, 50%, 38% and 4% of length.

Fig. 2. pQCT scans (raw) of the left upper limb of a control participant at four different sites along the length of the humerus and forearm. A: 4% forearm; B: 50% forearm; C: 66% forearm and D: 50% humerus. The ulna is located on the left and the radius on the right in images A-C. Cortical bone is displayed in (white), muscle (light grey), adipose tissue (dark grey), and air (black).

Fig. 3. pQCT scans (raw) of the left lower leg of a control participant at three different sites along the limb segment. A: 4% lower leg; B: 38% lower leg; C: 50% lower leg. The fibula is located on the left and the tibia on the right in each of the three images. Cortical bone is displayed in (white), muscle (light grey), adipose tissue (dark grey), and air (black).

Fig. 4. Comparisons of mean cortical areas (*CA*) along the upper limb for swimmers, cricketers and controls: (A) midshaft humerus (50%); (B) mid-proximal radius (66%); (C) midshaft radius (50%); (D) distal radius (4%). All values have been standardized. Boxes represent the 25th–75th percentile range, with the horizontal dark line indicating the median, and whiskers extending to the maximum and minimum values within 1.5 box lengths. Outliers are indicated with an o.

Fig. 5. pQCT scans of the left lower leg midshaft (50% of length), mid-distal (38%) and distal (4%) sections of a hockey player and distance runner displaying extreme examples of cortical bone distribution and thickness.