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Torsional stiffness and strength of the proximal tibia are better predicted by finite element models than DXA or QCT

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ABSTRACT

Individuals with spinal cord injury experience a rapid loss of bone mineral below the neurological lesion. The clinical consequence of this bone loss is a high rate of fracture around regions of the knee. The ability to predict the mechanical competence of bones at this location may serve as an important clinical tool to assess fracture risk in the spinal cord injury population. The purpose of this study was to develop, and statistically compare, non-invasive methods to predict torsional stiffness (K) and strength (T_{ult}) of the proximal tibia. Twenty-two human tibiae were assigned to either a "training set" or a "test set" (11 specimens each) and mechanically loaded to failure. The training set was used to develop subject-specific finite element (FE) models, and statistical models based on dual energy x-ray absorptiometry (DXA) and quantitative computed tomography (QCT), to predict K and T_{ult}; the test set was used for cross-validation. Mechanical testing produced clinically relevant spiral fractures in all specimens. All methods were accurate and reliable predictors of K (cross-validation $r^2 \geq 0.91$; error$\leq$13%), however FE models explained an additional 15% of the variance in measured T_{ult} and illustrated 12–16% less error than DXA and QCT models. Given the strong correlations between measured and FE predicted K (cross-validation $r^2 = 0.95$; error = 10%) and T_{ult} (cross-validation $r^2 = 0.91$; error = 9%), we believe the FE modeling procedure has reached a level of accuracy necessary to answer clinically relevant questions.

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1. Introduction

Individuals with spinal cord injury (SCI) experience a rapid loss of bone mineral at regions below the neurological lesion. Depending on the anatomic location, some 25% to 50% of their lower-extremity bone mineral is resorbed within the first 2 to 3 years of SCI (Biering-Sorensen et al., 1990; Eser et al., 2004). The clinical consequence of this reduction in bone is an increased lifetime risk of low-energy fracture that is two times greater than the general population (Vestergaard et al., 1998). These fractures are a source of considerable morbidity; more than 50% are characterized by medical complications requiring prolonged hospitalization (Morse et al., 2009).

Fractures in individuals with SCI frequently occur around regions of the knee, e.g. the proximal tibia (Comarr et al., 1962; Eser et al., 2005; Freehafer et al., 1981; Morse et al., 2009; Zehnder et al., 2004). Common causes of fracture include transfers, falls from wheelchairs, and rolling over in bed (Comarr et al., 1962; Eser et al., 2005; Freehafer et al., 1981; Morse et al., 2009; Zehnder et al., 2004). Torsional loading has been implicated as a principal mode of failure, as spiral fracture patterns are frequently observed in this population (Keating et al., 1992; Martínez et al., 2002). The current fracture risk assessment tools for the general public are inadequate for people with SCI. In part, this is because the locations of routine fracture do not correspond between these two groups. Therefore, the ability to quantify the mechanical competence of bone at a physiologically relevant location may serve as an important clinical tool to assess fracture risk in the SCI population.

The purpose of this study was to develop and statistically evaluate three non-invasive methods for predicting torsional stiffness and strength of the proximal tibia. These methods included (1) subject-specific finite element (FE) models, and multivariate models based on (2) dual energy x-ray absorptiometry (DXA) and (3) quantitative computed tomography (QCT). Because the FE method explicitly models structural and material behavior, while DXA and QCT derived predictions are inherently statistical, we hypothesized that FE models would be a more accurate and reliable predictor of mechanical behavior.

2. Methods

2.1. Specimens

Twenty-two tibiae were excised from formalin-fixed cadavers of mixed death histories (ages 46–98 years, 11 females, 17 right limbs). All specimens were free

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from overt structural pathology as identified by an expert anatomist. Specimens were cleaned of soft tissue and osteotomy was performed 15 cm distal to the intercondylar eminence. The proximal and distal most 2 cm of bone were potted in polymethyl methacrylate (PMMA), leaving 11 cm of bone exposed. Specimens were assigned to either a "training" set or a "test" set (11 specimens each) based on DXA assessed areal bone mineral density (aBMD) of the entire proximal tibia (see Image acquisition and analysis). Sets were allocated by ranking aBMD of the entire sample and assigning every other ranked specimen to a specific set.

2.2. Image acquisition and analysis

The DXA and computed tomography (CT) data were acquired with the specimens aligned along the longitudinal axis and fully immersed in water. The DXA scans were performed using a Hologic QDR-4500 (Holigic, Waltham, MA) with the lumbar spine acquisition software. Three regions of bone were analyzed corresponding to 0–10%, 10–20%, and 20–30% of tibial length (medial condyle to medial malleolus), as measured from the proximal end. These regions, which illustrated considerable variation in their trabecular and cortical bone makeup, were chosen based on their anatomical correspondence to epiphyseal (0–10%), metaphyseal (10–20%), and diaphyseal (20–30%) locations (Fig. 1). For each region, bone mineral content (BMC), and aBMD were computed.

The CT scans were performed using a BrightSpeed (GE Medical Systems, Milwaukee, WI) with acquisition settings of 120 kV, 200 mA, an in-plane resolution of 0.352 mm, and a slice thickness of 0.625 mm. Each scan included a calibration phantom (QRM, Moehrendorf, Germany) with known bone equivalent concentrations. The phantom was used to convert CT attenuation in Hounsfield units (Hu) to calcium hydroxyapatite equivalent density (ρ_{ha}).

The QCT analysis was performed using a combination of Mimics (Materialise, Leuven, Belgium) and Matlab (MathWorks, Natick, MA) software. Measurements were computed for epiphyseal, metaphyseal, and diaphyseal regions as defined above (Fig. 1). Proximal tibiae were segmented using a 0.15 g/cm^3 threshold to identify the periosteal surface boundary. For the epiphyseal and metaphyseal regions, trabecular volumetric BMD (Tb.vBMD) and cortical BMC (Ct.BMC) were computed. Trabecular and cortical bone specific regions were identified using methods similar to those described for the proximal femur by Lang et al. (2004). The trabecular region was determined from a 3.5 mm, or 10 pixel, in-plane erosion of the integral bone region (i.e., all voxels contained within the periosteal surface boundary). The cortical region was determined from a Boolean subtraction of the trabecular from the integral region, followed by a thresholding of 0.35 g/cm^3 to remove any residual trabecular bone (Fig. 1). For the diaphyseal region, only cortical BMC was computed. Here, the cortical region included all voxels within the integral region greater than 0.35 g/cm^3. Measures of geometry and strength were also computed for each region including cortical bone volume (Ct.BV), integral bone cross sectional area (CSA), and an integral bone torsional strength index (TSI) (see Supplementary material).

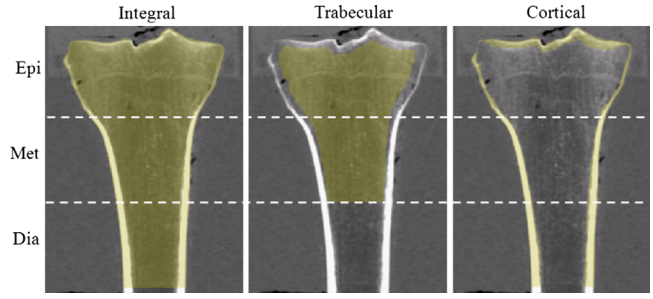

Fig. 1. A frontal plane cut-through view of the QCT regions (Epi, Met, Dia) of interest for a representative specimen. Yellow pixels superimposed on the image correspond to integral, trabecular and cortical regions of bone. (For interpretation of the references to color in this figure caption, the reader is referred to the web version of this article.)

2.3. Mechanical testing

Proximal tibiae were loaded in internal rotation using a materials testing machine (858 Mini Bionix II, MTS, Inc., Minneapolis, MN) with a custom linear actuated torsional device (Fig. 2a). The device has an experimental error less than 0.2 Nm (Edwards and Troy, 2012). Following 10 preconditioning cycles to 20 Nm, tibiae were loaded in internal rotation at a fixed rotation rate of 9.0 °/s until

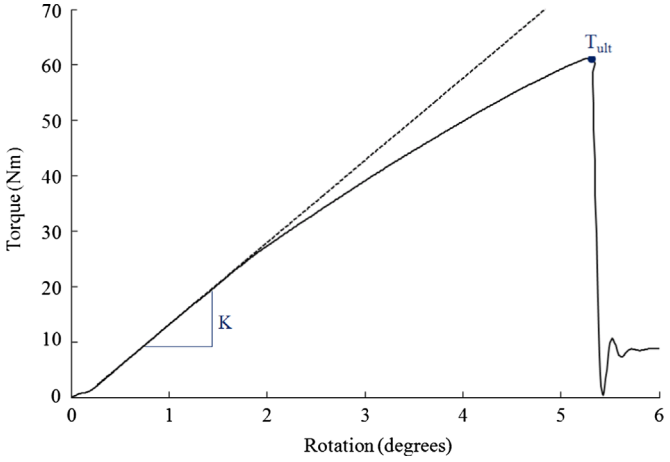

Fig. 3. Representative torque–rotation curve (solid line) illustrating the linear elastic projection (dashed line) used to calculate stiffness K, and the point on the curve corresponding to ultimate strength T_{ult}.

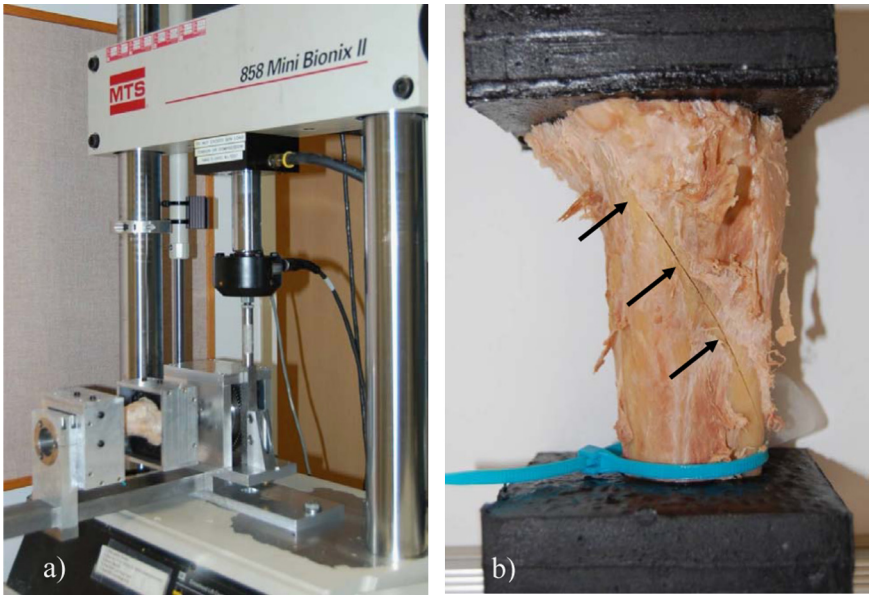

Fig. 2. (a) Proximal tibia being loaded in internal rotation using the linear actuated torsional device. (b) Lateral view of a proximal tibia illustrating a spiral fracture pattern (see arrows).

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