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Proximal femur bone strength estimated by a computationally fast finite element analysis in a sideways fall configuration

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ABSTRACT

Finite element (FE) analysis based on quantitative computed tomography (QCT) images is an emerging tool to estimate bone strength in a specific patient or specimen; however, it is limited by the computational power required and the associated time required to generate and solve the models. Thus, our objective was to develop a fast, validated method to estimate whole bone structural stiffness and failure load in addition to a sensitivity analysis of varying boundary conditions. We performed QCT scans on twenty fresh-frozen proximal femurs (age: 77±13 years) and mechanically tested the femurs in a configuration that simulated a sideways fall on the hip. We used custom software to generate the FE models with boundary conditions corresponding to the mechanical tests and solved the linear models to estimate bone structural stiffness and estimated failure load. For the sensitivity analysis, we varied the internal rotation angle of the femoral neck from −30° to 45° at 15° intervals and estimated structural stiffness at each angle. We found both the FE estimates of structural stiffness ($R^2=0.89$, $p<0.01$) and failure load ($R^2=0.81$, $p<0.01$) to be in high agreement with the values found by mechanical testing. An important advantage of these methods was that the models of approximately 500,000 elements took less than 11 min to solve using a standard desktop workstation. In this study we developed and validated a method to quickly and accurately estimate proximal femur structural stiffness and failure load using QCT-driven FE methods.

1. Introduction

While some hip fractures occur during normal activity (Parker and Twemlow, 1997), the impact forces encountered during a sideways fall are believed to be the main factor in over 90% of hip fractures (Grisso et al., 1991; Nevitt et al., 1989). Areal bone mineral density (aBMD) measured by dual energy x-ray absorptiometry (DXA) has been shown to be a significant predictor of fracture risk (Kanis, 2002) on a population level; however, 50% of fractures occur in people who, according to the World Health Organization aBMD criteria, would not be classified as osteopenic let alone as osteoporotic (Stone et al., 2003). Incorporating quantitative computed tomography (QCT) images and finite element (FE) analysis to estimate bone strength on a per-subject basis is a promising technique to improve the individual assessment of fracture risk. This noninvasive, subject-specific technique overcomes some of the inherent limitations of DXA (Bolotin, 2007) by incorporating volumetric measurements of density with structural information.

Finite element analysis and QCT images have been used to estimate whole bone stiffness and failure load in both standing and falling load configurations (Bessho et al., 2009, 2007; Cody et al., 1999; Dragomir-Daescu et al., 2011; Duchemin et al., 2008; Keyak et al., 1998, 2001; Koivumäki et al., 2012; Lochmüller et al., 2002; Lotz et al., 1991a, 1991b). These models have been in very good agreement with experimental mechanical testing of cadaver specimens (Bessho et al., 2007; Cody et al., 1999; Dragomir-Daescu et al., 2011; Duchemin et al., 2008; Keyak et al., 1998; Koivumäki et al., 2012) and have better agreement with femoral strength

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compared with aBMD by DXA or volumetric BMD by QCT (Cody et al., 1999). One of the main limitations of these models is their time-consuming and computationally expensive nature making them difficult to apply to clinical studies. Linear FE models have taken up to 6–8 h (Cody et al., 1999) while non-linear models can take approximately 10 h each (Bessho et al., 2009; Keyak, 2001).

The loading configuration of the proximal femur has a large influence on the experimentally measured (Pinilla et al., 1996) and estimated whole bone strength (Bessho et al., 2009; Keyak et al., 2001; Wakao et al., 2009). Pinilla and colleagues found that a 30° change in loading angle decreased the failure load by 24% (Pinilla et al., 1996). Since it is unknown how a specific person will fall (i.e. directly sideways, slightly turned forward or backward), and the large effect of loading direction on structural stiffness is well documented, it is important to determine the sensitivity of estimates of bone strength to various loading configurations. Again, applying this to a clinical population is limited by the computational time and power required to solve the models in various configurations. Another challenge with developing proximal femur FE models is the segmentation of the cortical and trabecular regions. In some regions, the thickness of the cortical shell can approach the resolution of the QCT system, leading to partial volume effects that can cause errors in the segmentation. An accurate segmentation is required to assign appropriate material properties to the cortical and trabecular regions (Hangartner, 2007) and for accurate post-processing to determine the contributions of the cortical and trabecular regions to estimated whole bone strength.

Therefore, our primary objective was to develop a fast, validated method to estimate bone stiffness and failure load. Our secondary objective was to determine the sensitivity of boundary conditions and the contributions of the cortical and trabecular regions to the overall bone stiffness.

2. Materials and methods

2.1. Specimens

We collected twenty fresh-frozen proximal femurs (5 male, 15 female; average age: 77±13 years) from the Gross Anatomy Lab at the University of Calgary. No specific requirements were used for the selection and no medical history was available so it was not known if previous bone diseases were present. Before scanning and mechanical testing, we cleaned the specimens of soft tissue, kept them frozen in saline soaked gauze, and thawed them overnight. The Conjoint Health Research Ethics Board at the University of Calgary approved all procedures.

2.2. CT scan acquisition

Prior to scanning, all specimens were secured in saline filled tube in order to simulate the attenuation of soft tissue. The tubes were placed on top of a hydroxyapatite calibration phantom (B-MAS200, Kyoto Kagaku, Japan) that contained five hydroxyapatite rods (0, 50, 100, 150, 200 mg/cm^3) and extended the length of the scan region. We scanned all specimens with CT (GE Discovery CT750HD, GE Healthcare; 120 kVp, 60 mAs, 512×512 matrix size) and reconstructed the images with a slice thickness of 0.625 mm and an in-plane resolution of 0.439 mm×0.439 mm. Femoral neck (FN) aBMD by DXA (Hologic QDR4500, Hologic; Bedford, MA) was also measured for all specimens.

2.3. Mechanical testing

We performed the mechanical testing in a configuration designed to simulate a sideways fall on the hip (de Bakker et al., 2009; Courtney et al., 1994; Eckstein et al., 2004; Manske et al., 2006; Pulkkinen et al., 2006). The testing apparatus positioned the femoral shaft at an angle of 10° from horizontal and the femoral neck internally rotated to 15°. The femoral shaft was free to translate in the sagittal plane and free to rotate around the anterior-posterior axis at the distal end. The femoral head was also free to translate in the sagittal plane but constrained in the direction of loading. The femoral head and the greater trochanter were both embedded in polymethylmethacrylate (PMMA; Fastray, Bosworth Company, Skokie, Il.) caps to prevent local crushing and improve load distribution.

The mechanical testing apparatus and specimen was placed within a materials testing system (Instron 8874; Instron Corp; Canton, MA) with a 25 kN rated load cell (Sensor Data M211-11; Sterling Heights, MI) (Fig. 1). A data acquisition card (National Instruments PCI-6040E; Austin TX) signal conditioned with a 10 kHz hardware cutoff filter and sampled at 20 kHz recorded the load and displacement. We used a 2000 N/V output resulting in a bit level resolution of 4.9 N/bit for the load and 5 mm/V resulting in a 0.012 mm/bit resolution for the displacement. First, we applied a preload of 100 N followed by a hold of 2 s. Next we applied a constant displacement at a rate of 2 mm/s (Courtney et al., 1994). From the force-displacement curve, we determined the stiffness based on the slope of the linear region of the curve and failure load as the maximal load encountered during the test (Pistoia et al., 2002).

2.4. Image processing

To find the periosteal and endosteal surfaces we used a semi-automatic contouring method (Stradwin 4.3; Cambridge, UK) (Treece et al., 2010). This software produces an unbiased cortical thickness measurement to 0.3 mm (Treece et al., 2010) and also outputs the cortical and trabecular surfaces. The surfaces were used to classify the cortical and trabecular regions to the nearest voxel. In-house software developed using the Visualization Toolkit (VTK 5.6; Kitware Inc.; Clifton Park, NY) calibrated the QCT measured density values based on the calibration phantom and rescaled the images using cubic interpolation, as previously suggested (McErlain et al., 2012), to 1.0 mm isotropic voxels.

2.5. Finite element analysis

To create the FE model, we converted each of the voxels in the image to an 8-node hexahedral element and assigned a Poisson's ratio of 0.3 (Varghese et al., 2011). The calibrated density values (ρ_{ash}, mg/cm^3) were converted to Young's modulus values (E, GPa) using the relationship found by Keller (Keller, 1994):

$$E = 10.50\rho_{ash}^{2.29} \quad (1)$$

The boundary conditions in the FE model mimicked the mechanical testing conditions that represent a sideways fall on the hip. To apply these boundary conditions, an operator first aligns the femoral neck vertically and the femoral shaft horizontally. The software (v5.4, Numerics88 Solutions; Calgary, Canada and VTK 5.6; Kitware Inc.; Clifton Park, NY) automatically transforms the image to the desired loading configuration and detects the femoral head, trochanter, and femoral shaft surfaces. The surface nodes where the forces were applied were matched to the surface applied in the mechanical tests based on the depth of the PMMA caps on both the femoral head and the greater trochanter. These same procedures can be implemented for *in vivo* images where the forces are automatically applied to a contact surface of nodes without further operator intervention. We assigned elements representing the PMMA caps a Young's modulus of 2500 MPa and a Poisson's ratio of 0.3 (Lewis, 1997) and fixed the surface nodes on the femoral head PMMA cap in the loading direction, but they were free to move in

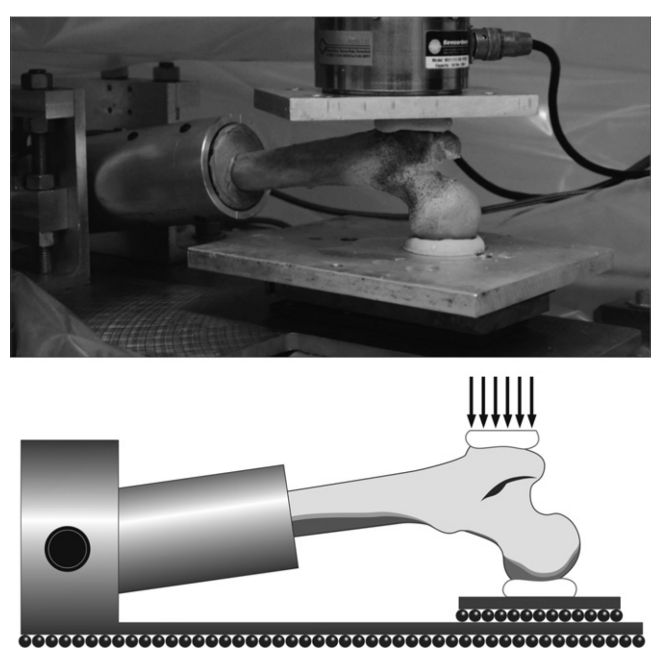

Fig. 1. Photograph of the testing apparatus and specimen positioned in the mechanical testing device (top) and corresponding schematic of the finite element model boundary conditions where a 1 mm displacement is applied to the PMMA cap (represented in white) on the greater trochanter (bottom).