Bone 50 (2012) 824–829

Contents lists available at SciVerse ScienceDirect

Bone

journal homepage: www.elsevier.com/locate/bone

Original Full Length Article

Ct-based finite element models can be used to estimate experimentally measured failure loads in the proximal femur

Janne E.M. Koivumäki [a,*], Jérôme Thevenot [a], Pasi Pulkkinen [a], Volker Kuhn [b,c], Thomas M. Link [d], Felix Eckstein [c,e], Timo Jämsä [a,f]

[a] Department of Medical Technology, University of Oulu, P.O. Box 5000, FI-90014 Oulu, Finland
[b] Medical University, Innsbruck, Austria
[c] LMU München, München, Germany
[d] University of California, San Francisco, CA, USA
[e] Paracelcus Medical University, Salzburg, Austria
[f] Department of Diagnostic Radiology, Oulu University Hospital, Oulu, Finland

ARTICLE INFO

Article history:
Received 8 December 2011
Revised 17 January 2012
Accepted 18 January 2012
Available online 28 January 2012

Edited by: Thomas Einhorn

Keywords:
Hip fracture
Finite element
Bone biomechanics
Bone strength
Femur

ABSTRACT

The objective of this experimental finite element (FE) study was to assess the accuracy of a simulation model estimate of the experimentally measured fracture load of the proximal femur in a sideways fall. Sixty-one formalin-fixed cadaver femora (41 female and 20 male) aged 55–100 years (an average of 80 years) were scanned with a multi-detector CT scanner and were mechanically tested for failure in a sideways fall loading configuration. Twenty-one of these femurs were used for training purposes, and 40 femurs were used for validation purposes. The training set FE models were used to establish the strain threshold for the element failure criteria. Bi-linear elastoplastic FE analysis was performed based on the CT images. The validation set was used to estimate the fracture loads. The Drucker-Prager criterion was applied to determine the yielding and the maximum principal stress criteria and the minimum principal strain criteria for element failure in tension and in compression, respectively. The estimated fracture load values were highly correlated with the experimental data ($r = 0.931$; $p < 0.001$). The slope was 0.929, with an intercept of 258 N, which was not significantly different from 1 and 0, respectively. The study shows that it is possible to estimate the fracture load with relatively high accuracy in a sideways fall configuration by using the CT-based FE method. This method may therefore be applied for studying the biomechanical mechanisms of hip fractures.

© 2012 Elsevier Inc. All rights reserved.

Introduction

Sideways falls to the greater trochanter are a significant cause of hip fracture and are frequently associated with osteoporosis. An osteoporotic fracture is usually related to reduced bone strength and falling. In individuals over 65 years of age, falls are the main etiological factor in over 90% of hip fractures, and a history of falls is associated with an increased risk of hip fractures [1,2]. Studies on the failure mechanism of hip fracture in the sideways fall loading configuration might offer improved tools for individual fracture risk assessment and focused preventative actions.

Computational finite element (FE) analysis is a useful method for studying the mechanical characteristics of hip fracture. Previously, FE analysis has been recognized as a noninvasive tool to estimate fracture load [3–16] or the risk for a specific fracture type [3,17–20]. CT-based nonlinear FE analysis that incorporates three-dimensional geometry and bone density distribution has been shown to produce an estimate of the fracture load of the proximal femur with reasonable accuracy for given boundary conditions [4,9]. This analysis usually requires high computational power and is time consuming. However, before a simple, straightforward model can be produced for clinical use, accurate models are required as a reference for reduced, more simplified representations.

Bessho et al. and Keyak [4,9] used a stance loading configuration, which may not adequately elucidate the failure mechanisms behind the clinical osteoporotic hip fractures that typically occur in sideways falls. There are only a few previous FE studies that estimate the experimental fracture load in a configuration that simulates a fall to the side [7,12–14]. However, only a limited amount of femur specimens or linear FE analyses were used in these studies. The generation of an accurate FE model using nonlinear analysis with a larger sample size for a sideways fall configuration is therefore required. This model may also contribute to further studies, e.g., when the fracture mechanisms behind a fall are studied in various loading configurations. The objective of this experimental FE study was to assess the accuracy of a simulation model that estimates the experimentally

* Corresponding author. Fax: +358 8 537 6000.
E-mail address: janne.koivumaki@oulu.fi (J.E.M. Koivumäki).

8756-3282/$ – see front matter © 2012 Elsevier Inc. All rights reserved.
doi:10.1016/j.bone.2012.01.012

measured fracture load of the proximal femur in a sideways loading configuration.

Materials and methods

Study sample

Cadaver femora aged 55–100 years (an average of 80 years) from a larger experimental study [21,22] were scanned with a multi-detector computed tomography (MD-CT) scanner and mechanically tested for failure in a sideways fall loading configuration, as described previously [21]. Sixty-one femurs (41 female and 20 male) were randomly selected for nonlinear FE analysis to estimate the fracture loads. Twenty-one of these femurs (16 female and 5 male) were used for training, and 40 femurs (25 female and 15 male) were used for validation purposes. The characteristics of the selected femurs are represented in Table 1.

The specimens were obtained from the Institute of Anatomy at the Ludwig Maximilians University of Munich (Germany) and were fixed in alcohol/formalin solution. The sample preparation and storage methods have previously been presented in detail [21]. Individuals with bone diseases other than osteoporosis or osteopenia were excluded from the study, based on biopsies of the specimens.

In vitro dual X-ray absorptiometry (DXA) scans of the femora were obtained with a standard narrow fan beam scanner (GE Lunar Prodigy, GE Lunar Corp., Madison, WI, USA) with the femoral specimens submerged in a water bath. Standard positioning was used across all specimens, and the proximal femoral bone mineral density (BMD) and related T-score values were evaluated with the software provided by the manufacturer.

The mechanical testing setup has been described previously [21,22]. In brief, the femora were tested in a side-impact configuration in which the femoral shaft was positioned 10° from horizontal, the neck was positioned at a 15° internal rotation, and the femoral head faced downward. The length of the femoral shaft above the fixation on the shaft was four times the head diameter. Two halves of a tennis ball were used in the femoral head. The loads were applied to the greater trochanter through a pad. The load was applied at a rate of 6.6 mm/s. The failure load (the maximal load encountered during the test) was determined from the load–deformation curve.

MD-CT scans

The MD-CT scan setup has been presented previously [19]. Briefly, the specimens were degassed, packed in airproof plastic bags filled within a formalin/water solution, and positioned in the scanner (Sensation 16; Siemens Medical Solutions, Erlangen, Germany) in a position that was comparable to that used in the in vivo exam of the pelvis and proximal femur. A slice thickness of 0.75 mm was used. The settings were 120 kVp and 100 mAs, a 512 × 512 pixels image matrix, and a field of view of 100 mm. The in-plane spatial resolution was 0.25 mm × 0.25 mm. A reference phantom (Osteo Phantom, Siemens, Erlangen, Germany) was placed below the specimens. The hydroxyapatite phantom was composed of 0 mg/cm^3 and 200 mg/cm^3 elements representing the water-like and bone-like parts of the phantom, respectively.

Table 1
Study sample characteristics. The total FE sample set consisted of 61 cadaver femora. Twenty-one femora were randomly selected for the training set, and 40 femora were randomly selected for the validation set.

	All (n = 61)	Training set (n = 21)	Validation set (n = 40)	p
Females	41	16	25	
Males	20	5	15	
Age (years)	80.4 ± 9.2	80.3 ± 10.4	80.5 ± 8.6	0.959
Height (cm)	161.3 ± 8.0	158.8 ± 8.4	162.8 ± 7.5	0.067
Weight (kg)	55.9 ± 12.9	49.1 ± 11.8	59.5 ± 12.2	0.003
BMI (kg/m^2)	20.9 ± 4.0	19.6 ± 4.7	21.7 ± 3.4	0.054
Proximal femoral BMD (g/cm^2)	0.72 ± 0.15	0.70 ± 0.13	0.75 ± 0.17	0.203
Proximal femoral T-score	−2.63 ± 1.16	−2.77 ± 0.99	−2.53 ± 1.27	0.533

Values are given as the mean ± SD. The p-values for the difference between the training set and validation set (independent samples T-test).
BMI = body mass index. BMD = bone mineral density.

FE models

Proximal femurs were segmented with MD-CT datasets and Mimics (v12.1, Materialise, Leuven, Belgium). Trabecular and cortical bone were modeled individually based on the study of Chevalier et al., in which the use of an explicit cortex in the FE models increased the correlations to the experimental stiffness and strength [23]. Previous studies that investigated the model convergence recommended using an element size of 3 mm or smaller to illustrate the sharp variations in the mechanical properties and to avoid a decrease of the predicted stresses and strains that are evident with larger elements [4,24]. To test the convergence, we created models from the training set with maximum element sizes of 2 mm, 3 mm and 4 mm. The model geometry remained constant regardless of the element size. The total strain energy and the computing time were compared between the models under the same boundary conditions and a load of 2000 N [4] (Table 2). To compromise between the model accuracy and the computing time, trabecular and cortical bone were modeled with 10-noded tetrahedral elements with a maximum element size of 3 mm and a minimum cortical thickness of 1 mm. Femap (Femap, version 10.1., UGS Corp., Plano, TX, USA) software was used for the pre-processing, and MD Patran (2008 r1, MSC.Software Corporation, Santa Ana, USA) software was used for the post-processing of the models. MD Nastran (R3, MSC.Software Corporation, Santa Ana, USA) was used for the calculation.

For material heterogeneity, an average gray value of all of the voxels inside an element was calculated in Mimics. The bone equivalent density (ash density, ρ_{ash}) was then defined by assuming a linear relationship in which the density is proportional to the attenuation. The Young's modulus (E, MPa) was estimated by using the equation

$$E = 10.095\,\rho \frac{\text{MPa}}{\text{mg/cm}^3} \tag{1}$$

in which ρ is the equivalent CT density (mg/cm^3) [25]. The subject-specific maximum density value of the cortical bone was used to obtain the isotropic Young's modulus for the simulated cortical bone. Trabecular bone was considered to be a heterogeneous isotropic material, and the subject-specific material properties were assigned to each trabecular bone element. A Poisson's ratio of 0.33 was selected for the models [26]. The Drucker-Prager yield criterion [27] was used as previously suggested [4]; this criterion is described by the following equations:

$$F(\sigma) = 3\alpha\sigma_m + \sqrt{J_2'} \tag{2}$$

$$J_2' = \frac{1}{2}\left(\sigma_x'^2 + \sigma_y'^2 + \sigma_z'^2\right) + \tau_{xy}^2 + \tau_{yz}^2 + \tau_{xz}^2 \tag{3}$$

Table 2
The effect of the maximum element size on the total strain energy and the computing time for the models with the same boundary conditions and a load of 2000 N. A convergence test with two samples.

	Maximum element size		
	4 mm	3 mm	2 mm
Total strain energy (%)	104–107%	102–104%	100%
Computing time	100%	300%	700%

Download English Version:

https://daneshyari.com/en/article/2779564

Download Persian Version:

https://daneshyari.com/article/2779564

Daneshyari.com