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To what extent can linear finite element models of human femora predict failure under stance and fall loading configurations?

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ABSTRACT

Proximal femur strength estimates from computed tomography (CT)-based finite element (FE) models are finding clinical application. Published models reached a high in-vitro accuracy, yet many of them rely on nonlinear methodologies or internal best-fitting of parameters. The aim of the present study is to verify to what extent a linear FE modelling procedure, fully based on independently determined parameters, can predict the failure characteristics of the proximal femur in stance and sideways fall loading configurations.

Fourteen fresh-frozen cadaver femora were CT-scanned. Seven femora were tested to failure in stance loading conditions, and seven in fall. Fracture was monitored with high-speed videos. Linear FE models were built from CT images according to a procedure already validated in the prediction of strains. An asymmetric maximum principal strain criterion (0.73% tensile, 1.04% compressive limit) was used to define a node-based risk factor (RF). FE-predicted failure load, mode (tensile/compressive) and location were determined from the first node reaching RF=1.

FE-predicted and measured failure loads were highly correlated (R^2=0.89, SEE=814 N). In all specimens, FE models correctly identified the failure mode (tensile in stance, compressive in fall) and the femoral region where fracture started (supero-lateral neck aspect). The location of failure onset was accurately predicted in eight specimens.

In summary, a simple FE model, adaptable in the future to multiple loads (e.g. including muscles), was highly correlated with experimental failure in two loading conditions on specimens ranging from normal to osteoporotic. Thus, it can be suitable for use in clinical studies.

1. Introduction

The heterogeneity of modelling approaches, and results, among the few finite element (FE) studies addressing proximal femur fracture in clinical cohorts (Keyak et al., 2011, 2013; Orwoll et al., 2009; Nishiyama et al., 2014), indicates that no consensus has yet been reached about how to obtain bone strength estimates from subject-specific FE models based on computed tomography (CT) images.

The importance of a thorough in-vitro validation of modelling procedures before they are applied to clinical studies is well acknowledged (Viceconti et al., 2005), and led to the publication of numerous validation studies in the past decade. Published models reached notable accuracy levels (determination coefficient R^2 ranging from 0.73 to 0.96, Standard Error of the Estimate SEE ranging from 228 to 1560 N) in predicting the failure load of excised proximal femora under different loading conditions: (i) quasi-axial loads mimicking the direction of hip joint reaction in physiological activities (also known for short as “stance”) (Bessho et al., 2007; Duchemin et al., 2008; Keyak et al., 2005); (ii) forces simulating an unprotected fall to the side (“fall”) (Dragomir-Daescu et al., 2011; Koivumäki et al., 2012; Nishiyama et al., 2013). Only two works (Dall’Ara et al., 2013; Keyak et al., 1998) studied both configurations.

Many of the models so far proposed rely on quite complex methodologies, that imply defining material non-linearities (Bessho et al., 2007; Dall’Ara et al., 2013; Dragomir-Daescu et al., 2011; Keyak et al., 2005; Koivumäki et al., 2012) or geometric ones (Duchemin et al., 2008). Often, model parameters were not taken from independent sources but determined internally through best fitting in a training set of specimens, and then applied to a test set (Dragomir-Daescu et al., 2011; Keyak et al., 2005; Nishiyama et al., 2013).

Recently, we proposed a modelling procedure that in three specimens was able to identify proximal femur fracture patterns

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through linear analyses and a simple maximum principal strain criterion (Schileo et al., 2008a). The procedure is fully based on independently determined parameters and has been validated in terms of strain predictions in the elastic field in both stance (Schileo et al., 2007, 2008b) and fall configurations (Grassi et al., 2012).

The aim of the present study is to verify to what extent this simple FE modelling procedure can predict the failure characteristics of the proximal femur in both stance and sideways fall loading configurations.

2. Materials and methods

To perform the present FE validation study, we extracted, from all the experimental data regarding failure tests of proximal femora that were available in our laboratory, all bones for which CT images, specimen digitization, and well-conditioned failure tests were available. From a total of 31 bones (20 tested to failure in stance, 11 in fall), we excluded from the study: femora that presented problems in tissue-preservation: embalming (7), long and/or subsequent uncontrolled defrosting periods (3); femora on which we could not perform any validation of strain or displacements (2, which were never instrumented with strain gauges and/or displacement transducers); femora that showed failure patterns not physiologic and not compatible with a robust estimate with FE simulation (3 femora failed by trochanteric crushing within the trochanteric support); femora for which CT dataset was not available (1) or presented artefacts not compatible with an accurate model generation (1).

The 14 femora that were eligible for inclusion in the study belonged to several different experimental campaigns that spanned a course of more than four years. Consequently, some controlled differences in the experimental tests among bones are present, which simply reflect the evolution in the experimental set-up. Subsets of the experimental results on some of the bones included in the present study were already presented in other papers (Cristofolini et al., 2011; Juszczyk et al., 2011, 2013).

2.1. Specimen details and diagnostic assessments

Fourteen fresh-frozen cadaver femora were obtained (IIAM, Jessup, PA, US; and Anatomy Gifts Registry, Hanover, MD, US) (Table 1). Age was similar in stance and fall groups (mean 78 vs. 76, Mann-Whitney $p=0.9$). Bone density was not significantly different (mean densitometry T-score -3.3 vs. -1.7, $p=0.4$), but while stance specimens all came from osteopenic or osteoporotic donors, fall specimens spanned a wide range of bone densities. During the tests, specimens were wrapped in cloths soaked with physiological solution, and kept at $-25\,^{\circ}$C when not in use. All the specimens were CT-scanned with a clinical protocol, using the European Spine Phantom (ESP) (Kalender, 1992) for densitometric calibration.

2.2. Experimental tests

The experimental procedures followed published protocols and are here briefly summarised for clarity (Fig. 1). An anatomical reference system was identified on each femur according to Cristofolini and Viceconti (1999), Ruff and Hayes (1983). Femora were distally potted with bone cement to 66% of the femoral shaft length (specimens #1–3) or resected and potted to 33% of the femoral shaft length (#4–14).

2.2.1. Stance

Seven femora (specimens #1–7) were tested to failure in a stance configuration, according to Cristofolini et al. (2007), Juszczyk et al. (2011). The distal femur was medially tilted 8° in the frontal plane, and constrained. A vertical force was applied on the femoral head, through a system of linear bearings to eliminate any horizontal force component. A spherical cap of bone cement protected the femoral head. Grease was used to minimise friction at the cap–bone interface. Load was applied at a constant displacement rate of 2 (specimens #1–3) or 20 mm/s (#4–7), leading to an average strain-rate in the most stressed regions during the loading ramp around 5000 (#1–3) and 50,000 $\mu\varepsilon$/s (#4–7), and to bone fracture in 2–4 s (#1–3) and tenths of a second (#4–7).

2.2.2. Fall

Seven femora (specimens #8–14) were tested to failure in a sideways fall configuration, according to Grassi et al. (2012), Zani et al. (2013). The femora were tilted by 10° in the frontal plane, and internally rotated by 15°, similarly to most of the analogous tests reported in literature (Koivumäki et al., 2012; Nishiyama et al., 2013). They were constrained distally through a hinge that allowed tilt in a quasi-frontal plane. The greater trochanter rested on a flat surface, and load was applied vertically to the femoral head. A system of linear bearings eliminated any horizontal force component. An aluminium spherical cap applied with bone cement protected the greater trochanter and the femoral head. Grease was used to minimise friction at the cap–bone interface. The load actuator speed was tuned so that the average strain-rate in the most stressed regions during the loading ramp was around 50,000 $\mu\varepsilon$/s, consistent with reports on bone strain rate in strenuous activities (Al Nazer et al., 2012) in order to generate failure in about 0.2 s. The resulting actuator speed was between 17.5 and 32.5 mm/s (Table 1).

2.2.3. Measurements

Load–displacement curves were obtained from the loading machine (Mod. 8502, Instron, USA). Failure load was defined as the highest peak of the load–displacement curve. For three specimens (#4, 12 and 13) we took the peak load registered by the testing machine, because the load–displacement curve was not available.

A high-speed camera (Fastcam SA3, Photron, USA) operating at 3750–12,500 frames/s, was used to monitor the location of fracture onset. The camera monitored directly the supero-lateral or infero-medial aspect of the femoral neck, and indirectly (through two conveniently angled mirrors) the two adjacent anatomical aspects (Juszczyk et al., 2011). A pixel corresponded to approximately 0.2 mm on the physical specimen.

Table 1
Detailed donor data and brief summary of experimental loading conditions. The donors had no reported history of musculoskeletal disease. All specimens but #9 and #11 were unpaired.

All femora were CT-scanned (HiSpeed, GE Co., USA) with slice thickness 1 mm from femoral head to lesser trochanter, 5 mm elsewhere. Pixel size ranged from 0.48 to 0.66 mm. Peak voltage and tube current levels were 120 kVp and 160–180 mA, respectively.

Specimens #1–8 and #14 were examined with Dual-energy X-ray Absorptiometry (DXA) (Eclipse, Norland Co., USA). Areal bone mineral density estimated by DXA at the femoral neck is reported in terms of T-score, which is the number (with sign) of standard deviations from the mean value of young women.

Specimen no.	Donor						Loading		
	Side	Sex	Age	Height (cm)	Weight (N)	DEXA T-score	Config.	Rate (mm/s)	Constraint (% shaft) (%)
1	Right	Male	71	178	893	−1.87	Stance	2	66
2	Left	Male	73	175	716	−4.10		2	66
3	Left	Male	82	175	765	−4.09		2	66
4	Right	Male	77	175	824	−1.79		20	33
5	Right	Male	77	173	687	−3.60		20	33
6	Left	Female	80	160	1197	−3.80		20	33
7	Right	Male	83	175	824	−3.31		20	33
8	Right	Female	80	155	660	−4.07	Fall	32.5	33
9	Right	Female	84	168	630	−2.47*		17.5	33
10	Right	Male	62	173	1310	−3.81*		27.5	33
11	Left	Female	84	168	630	−1.30*		25	33
12	Right	Female	68	160	630	−2.48*		22.5	33
13	Right	Female	77	185	760	−3.69*		25	33
14	Left	Female	74	173	720	0.62		17.5	33

* For specimens #9–13 areal bone mineral density values were simulated from CT data, as in Keyak et al. (2011).