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Material mapping strategy to improve the predicted response of the proximal femur to a sideways fall impact

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ABSTRACT

Sideways falls are largely responsible for the highly prevalent osteoporotic hip fractures in today's society. These injuries are dynamic events, therefore dynamic FE models validated with dynamic *ex vivo* experiments provide a more realistic simulation than simple quasi-static analysis. Drop tower experiments using cadaveric specimens were used to identify the material mapping strategy that provided the most realistic mechanical response under impact loading. The present study tested the addition of compression-tension asymmetry, tensile bone damage, and cortical-specific strain rate dependency to the material mapping strategy of fifteen dynamic FE models of the proximal femur, and found improved correlations and reduced error for whole bone stiffness (R^2 = 0.54, RSME = 0.87 kN/mm) and absolute maximum force (R^2 = 0.56, RSME = 0.57 kN), and a high correlation in impulse response (R^2 = 0.82, RSME = 12.38 kg/s). Simulations using fully bonded nodes between the rigid bottom plate and PMMA cap supporting the femoral head had higher correlations and less error than simulations using a frictionless sliding at this contact surface. Strain rates over 100/s were observed in certain elements in the femoral neck and trochanter, indicating that additional research is required to better quantify the strain rate dependencies of both trabecular and cortical bone at these strain rates. These results represent the current benchmark in dynamic FE modeling of the proximal femur in sideways falls. Future work should also investigate improvements in experimental validation techniques by developing better displacement measurements and by enhancing the biofidelity of the impact loading wherever possible.

1. Introduction

Identifying individuals with a high risk of femoral fracture has proven difficult because current clinical measurements for osteoporosis, such as areal bone mineral density (aBMD) and FRAX scores, are unable to predict patient-specific fracture events resulting from low energy impact (Stone et al., 2003). To address this problem, finite element (FE) models of the proximal femur based on quantitative computed tomography (QCT) have been developed that can improve predictions of femoral strength, relative to femoral neck aBMD measurements *ex vivo* (Johannesdottir et al., 2017). Numerous studies have validated FE models with experimental testing of cadaver femurs in material testing machines and found high correlations between experimental and FE predicted femoral stiffness and fracture force (Grassi et al., 2012; Keyak, 2001). However, when these modeling techniques are applied to large clinical databases with fracture case history, the predictive capabilities were less powerful and improvements over conventional aBMD measurements were mild (Kopperdahl et al., 2014), suggesting that the models still lack important information to reliably differentiate fracture risk in the real world.

While contemporary modeling techniques are effective at capturing *ex vivo*, specimen-specific bone strength, these methods neglect the simulation of patient-specific loads. The first step in better understanding these loads is to improve modeling of the injury itself, which is the result of a dynamic event. While previously published experimental studies have tested a variety of material mapping methods and loading configurations, they customarily simulate quasi-static (QS) loading and are validated with QS *ex vivo* models. This neglects the physics of an

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actual impact event, which can be affected by properties such as viscoelasticity & viscoplasticity, inertia, material hardening, and shockwave propagation. Material testing systems generally apply displacements at 100 mm/s or lower, whereas impacts resulting during falls from standing height can occur at 3 m/s or higher (Feldman and Robinovitch, 2007). Furthermore, material testing systems typically ramp the displacement linearly to test the load bearing capacity of the femur, which differs from an impact test, in which the momentum is absorbed at a non-linear and non-constant deformation rate. Another critical difference is that the QS testing will inevitably fracture the bone, whereas a dynamic impact, e.g. with a drop tower test, delivers a specific energy, which does not necessarily result in a fracture. This could represent an important aspect of real-world fractures, considering that only 1–5% of falls actually result in fracture (Hayes et al., 1996; Nachreiner et al., 2007; Nevitt et al., 1989). Thus a dynamic FE model, that accurately represents the femur's response in fall-like impacts, could aid the identification of high risk patients by providing a more direct assessment of femur survivability.

Before dynamic models can be used to predict femur survivability in a sideways fall, they must first be validated with appropriate *ex vivo* experiments. In a previous study by our group, human cadaveric femurs were tested under both ramp-displacement and impact loading to compare differences in force-displacement behavior (Gilchrist et al., 2014; Ariza et al., 2015). It was shown that specimens under impact were stiffer but absorbed less energy than during QS testing. These experimental data were then compared to FE predicted whole bone stiffness, ultimate force, and energy using an explicit FE solver to simulate dynamic loading (Ariza et al., 2015). This initial attempt found weak correlations in stiffness and energy absorption, and no correlation in maximum force, which was attributed to a combination of pressure dependent yield behavior and tension-compression asymmetry not being modeled, and the lack of a clearly defined peak forces in the simulated force response.

The objective of the present study was to improve the mechanical behavior predicted by these dynamic FE models by redesigning the material mapping strategy, guided by observations from both the previous study, as well as the scientific literature. We applied only physically meaningful changes to the material description based on previous studies where direct mechanical testing was performed on bone tissue, with a preference for human bone studies where possible. We hypothesized that including tension/compression asymmetry, tensile bone damage, and more realistic strain rate dependent material properties would improve the accuracy of dynamic FE simulations of impact loading.

2. Materials and methods

2.1. Experimental testing

Details about the experimental drop tower testing have been published elsewhere (Gilchrist et al., 2013, 2014). Briefly, fifteen fresh frozen femurs (14 female/1 male, age: 76 ± 11 years, height: 160 ± 9 cm, Neck aBMD: 0.71 ± 0.11 g/cm^2) were scanned in high resolution-peripheral quantitative computed tomography (HR-pQCT) at 41 µm isotropic voxel resolution. The distal shaft of each femur was pinned between 290 and 305 mm from the proximal tip and potted in PMMA to a depth of 182 mm from the pin. Specimens were oriented with 15 degrees internal rotation and 10 degrees adduction (Courtney et al., 1994). Landmark locations were recorded relative to the experimental apparatus using an Optotrak Certus motion capture camera with a digitizer probe, RMSE = 0.1 mm (Rohling et al., 1995), which served to digitize the initial positon of the bone relative to the drop tower in order to match the alignment of the loading vector in the simulation to that of the experiment. The drop tower apparatus included a 32 kg impact mass, a 50 N/mm pelvis spring, a 2 kg pelvis mass, and 19 mm of closed cell foam to simulate soft tissue between the femur and the impact surface. A target impact speed of 3 m/s was applied, with displacement being tracked by high speed video recording at 9216 frames per second at a resolution of 575 × 288 pixels (5 pixels/mm). Force data were collected at 20 kHz from a +/− 13.34 kN (axial range), six-axis load cell.

For the present study, a qualitative analysis of the failure mode of each specimen was performed on the high speed video data from the experiments. The purpose was to link patterns in failure mode to over/under-prediction of the FE force-time response in the previous dynamic FE study (Ariza et al., 2015). These patterns were then used to guide the changes to the material mapping strategy in the present study. Five main fracture initiation sites were identified that encompassed all the fractures observed from the experiments, which included inter-trochanteric fracture, sub-capital cervical fracture, basi-cervical fracture, trochanteric crushing, and tensile failure of the inferior femoral neck (Supplementary Fig. 3).

2.2. Finite Element Models

The FE model construction has been previously described in detail elsewhere (Ariza et al., 2015). Briefly, the HR-pQCT images were first resampled to clinical-CT resolution in ITK-SNAP, and subsequently meshed with explicit, 10-node, parabolic, tetrahedral elements with minimum 3 mm target edge length in Ansys Workbench. The mesh was transformed to the experimental coordinate system using rigid registration with an iterative closest point algorithm. Boundary conditions mimicked the experimental set up (Fig. 1), with solid models representing PMMA supports molded to the bone surface, with a sliding frictionless contact between the bone and PMMA. A single node identified at the location of the distal pin was fully constrained except for rotation around the pin-axis. Above the greater trochanter PMMA pad, a rigid plate was modeled and assigned a velocity-time curve digitized from the displacement of a single pixel at the bone/PMMA interface that was tracked within the high speed video data with 0.2 mm resolution (Ariza et al., 2015). Controlling the simulation using the experimental velocity profile effectively isolated the effect of the material mapping strategy on the resulting force response. FE simulations were solved using the explicit LS-DYNA solver within ANSYS (v.14.5 ANSYS Inc., Canonsburg, USA).

The bottom PMMA cap on the femoral head was translationally constrained in the displacement axis, and fully rotationally constrained. While the bottom plate was translationally unconstrained in the experiment, it is unclear how much friction existed between the bottom PMMA cap and the steel plate, since slipping occurred in some cases when the femur began to break apart. In order to encompass the extreme possibilities, two model sets were constructed: in the first, the bottom nodes of the PMMA cap were left unconstrained, representing the no-friction scenario (Case I), while the second fully constrained the bottom nodes, representing the fully bonded scenario (Case II).

2.3. Material properties

2.3.1. Material model

Previous models used a piecewise_linear_plastic material (LS-DYNA Mat24) that was pressure independent to model bone tissue (Ariza et al., 2015). In the present study, both trabecular and cortical bone tissue were modeled using a rate-sensitive Fu-Change Foam (LS-DYNA Mat83), which uses constitutive equations developed by Chang (1995). This decision was motivated by recent findings that yielding and plateau behavior of trabecular bone in confined compression can be accurately simulated using crushable foam plasticity models (Kelly and McGarry, 2012). This specific material also enabled separate definitions of post-yield behavior and strain rate scaling between tension and compression, which was necessary for this study. Finally, element erosion was added to improve model convergence by assigning negligible stiffness to elements that result in a solver time step smaller than