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Importance of asymmetry and anisotropy in predicting cortical bone response and fracture using human body model femur in three-point bending and axial rotation

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ABSTRACT

Modeling of cortical bone response and failure is critical for the prediction of Crash Induced Injuries (CII) using advanced finite element (FE) Human Body Models (HBM). Although cortical bone is anisotropic and asymmetric in tension and compression, current HBM often utilize simple isotropic, symmetric, elastic-plastic constitutive models. In this study, a 50th percentile male femur FE model was used to quantify the effect of asymmetry and anisotropy in three-point bending and axial torsion. A complete set of cortical bone mechanical properties was identified from a literature review, and the femur model was used to investigate the importance of material asymmetry and anisotropy on the failure load/moment, failure displacement/rotation and fracture pattern. All models were able to predict failure load in bending, since this was dominated by the cortical bone material tensile response. However, only the orthotropic model was able to predict the torsional response and failure moment. Only the orthotropic model predicted the fracture initiation location and fracture pattern in bending, and the fracture initiation location in torsion; however, the anticipated spiral fracture pattern was not predicted by the models for torsional loading. The results demonstrated that asymmetry did not significantly improve the prediction capability, and that orthotropic material model with the identified material properties was able to predict the kinetics and kinematics for both three-point bending and axial torsion. This will help to provide an improved method for modeling hard tissue response and failure in full HBM.

1. Introduction and background

Human Body Models (HBMs) provide a computational platform to assess occupant response and the potential for injury in crash scenarios (Gierczycka et al., 2015; Schmitt et al., 2014), and, ultimately, to mitigate injury. An important goal for HBMs has been to predict injury at the tissue level and, in particular, to predict hard tissue failure or bone fracture, as this is often an injury of importance (AIS, 2005), or may be indicative of nearby soft tissue injuries (Tscherne and Oestern, 1982). However, the capability of HBMs to predict hard tissue fracture has been limited by the use of relatively simple constitutive models, for example isotropic metals plasticity models in the Global Human Body Models Consortium (GHBMC) and Total Human Model for Safety (THUMS) models, coupled with erosion-based material failure approaches (Hambli et al., 2012; Hambli and Thurner, 2013). The mechanical behavior of bone and the mechanism of bone fracture have been investigated using computational models at the microscopic and macroscopic levels for over 40 years (Zysset et al., 2013). Nevertheless, a widely accepted continuum treatment for cortical bone that can predict the failure load and fracture pattern for different modes of loading is not yet available.

Predicting fracture propagation and fracture pattern will enable HBMs to better predict the potential for injury in impact scenarios and could potentially enable HBMs to predict post-fracture response, which is acknowledged to present challenges in existing HBMs (DeWit and Cronin, 2012). In the short term, accurate prediction of cortical bone response and potential for failure is critical for HBM to address current challenges in human safety. Although mechanical testing of bone has identified asymmetry and anisotropy in bone material, the importance of these effects has not been assessed in full HBMs, where computational robustness and efficiency through a relatively coarse finite element mesh size are required.

This study aims to narrow that gap by identifying a set of cortical bone material properties from experimental studies and assessing the

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contribution of asymmetry and anisotropy to failure load and fracture pattern using a three-dimensional femur bone model from a current HBM assessed with independent whole-bone experimental data.

1.1. Mechanical response and modeling of bone

The complex hierarchical structure of bone has been associated with anisotropic material properties, asymmetry in compression and tension loading, and variations in properties with deformation rate (Cowin, 2001; Hansen et al., 2008; McElhaney, 1966; Reilly and Burstein, 1975; Turner, 2006). Although many reported properties exist for human and animal cortical bone, a single set of properties has not yet been proposed for general use for three-dimensional bone models in HBMs.

Many studies have investigated hard tissue constitutive models at the coupon level (Fondrk et al., 1999a, 1999b; Garcia et al., 2009, 2010; Iwamoto et al., 2005; Schileo et al., 2008; Tanaka et al., 2012; Zysset et al., 2013), but these approaches have not been translated to three-dimensional bone models in HBMs. The proposed constitutive models included a linear elastic-plastic formulation for low strain rates in the physiological range (Garcia et al., 2009, 2010; Tanaka et al., 2012), a non-linear visco-elastoplasticity formulation for a large strain deformation and higher strain rates (Fondrk et al., 1999a, 1999b), anisotropy (Iwamoto et al., 2005), a compression-tension asymmetry (Iwamoto et al., 2005; Niebur et al. 2000; Schileo et al., 2008), and a post-yield damage (Zysset et al., 2013). Two studies (Iwamoto et al., 2005; Martin et al., 1998) identified the bone strength asymmetry and anisotropy as the main contributors to bone fracture mechanisms and patterns. However, current methods of simulating cortical bone in HBMs treat the bone material as isotropic, with symmetric compression and tension properties modeled using metal plasticity approaches (e.g. Untaroiu et al., 2013).

Due to the complex geometry of bones and the need for computational efficiency in HBMs, continuum-level models with element sizes on the order of 1–3 mm are used. Asgharpour et al., (2014) extracted a femur from the THUMS HBM and developed a bone material model to replicate an experimental femur bending response; however, the mesh size dependency of the predicted response limited application of the material model proposed by Asgharpour et al., (2014) in other HBMs. In addition, strain rate dependent mechanical properties significantly over predicted the failure load 7 kN compared to an average failure force of 4.3 kN in the experiments (Funk et al., 2004) for dynamic three-point bending of the femur. In a similar study, Untaroiu et al. (2013) did not include rate effects in the material properties, since these properties were not consistently supported by the experimental data. He used an isotropic model with an effective plastic strain failure criterion but disabled the failure criterion in the area in which the impactor contacted the femur to avoid failure in the compression region, and predicted a failure force of 3.82 kN in three-point bending.

Model geometric simplifications have also been proposed, such as a two-dimensional (2D) femoral neck model (Hambli et al., 2012), with a clear limitation that it could only be applied to relatively simple loading, where a plane stress assumption incorporated in their model could not be applied to three-dimensional HBMs. Furthermore, material damage and crack propagation were implemented in a user-defined constitutive model, which is at present not available to other researchers (Hambli et al., 2012). In another study, Hambli and Thurner (2013) studied the quasi-static three-point bending load of a single trabeculae strut using a 2D finite element (FE) model with a user-defined material model. These methods involved a large computational cost, they are not generally available to the research community, and at present are not directly applicable in full HBMs with complex geometry.

1.1.1. Hard tissue failure

Modeling of hard tissue failure comprises two events: the accumulation of damage and initiation of the fracture, and fracture propagation. Micro-models with element sizes on the order of 1 μm have been used to investigate crack deflection into cement lines and resistance to crack propagation across osteons (Mischinski and Ural, 2011), which was found to decrease with increasing strain rate (Ural et al., 2011). Methods utilized to predict fracture at the continuum level include the cohesive zone model (CZM) (Ural et al. (2011), a reduced stiffness approach (Hambli and Thurner, 2013), an extended finite element (X-FEM) approach (Abdel-Wahab et al., 2012; Feerick et al., 2013; Li et al., 2013), and the element deletion or erosion approach (Hambli et al., 2012; Harrison et al., 2013). The main limitation of the CZM is a requirement to pre-define the fracture path for the analysis (Ali et al., 2014; Feerick et al., 2013).

Although the X-FEM approach is desirable to model crack propagation, it is computationally expensive and currently applicable in 2D models (Abdel-Wahab et al., 2012; Feerick et al., 2013; Hambli and Thurner, 2013; Li et al. , 2013). Furthermore, in a study that incorporated X-FEM in a 2D cortical bone model, simulation convergence and numerical stability presented challenges, due to multiple, mutually intersecting cracks predicted over a small region after a mesh refinement (Idkaidek and Jasiuk, 2016). Therefore, a smaller FE mesh requires a smaller time increment, leading to a longer simulation time, but without providing a guarantee of a more accurate result compared to a coarser mesh. Only a few studies (Ali et al., 2014; Giambini et al., 2016) have incorporated the X-FEM approach in 3D models, since this approach requires a custom solver and methods that are not commercially available at present.

While the element deletion (erosion) method is widely used to model material failure, one limitation is an aggregated treatment of crack initiation and propagation (Hambli et al., 2012, Hambli and Thurner, 2013). The fracture pattern is thus determined by the deleted elements of the mesh. In a more geometrically complex scenario, a study by DeWit and Cronin (2012) demonstrated that a relatively simple isotropic metal plasticity model with element erosion could predict the onset of failure, but generally did not predict the fracture pattern when a maximum plastic strain erosion criterion was used. The element erosion method has been shown to predict fracture patterns at the whole-bone level (Niebur et al., 2000; Schileo et al., 2008) for some modes of loading. A strain-based failure criterion is generally employed for numerical stability (DeWit and Cronin, 2012), and was justified by Nalla et al. (2003), who demonstrated experimentally that the local criterion for fracture in human cortical bone was consistent with a strain-based criterion used in theoretical and computational bone models.

1.2. Experimental data for model assessment

Evaluation of a computational model requires an objective assessment alongside the experimental data, in this case focused on cortical bone response and failure for whole-bone scenarios. The human femur is one of the most widely tested whole bones, and has been tested at the material level extensively, owing to the size of the bone and therefore the ability to extract test samples. Whole femur experimental tests include three-point bending, axial torsion and axial compression. Although axial compression loading is an important load case in vehicle crash scenarios due to interaction of the knee with the vehicle instrument panel, early investigations identified that the predicted response and failure were directly linked to the mesh transition between the diaphysis and epiphysis in the femur model. Therefore, an objective evaluation of the cortical bone material properties with this load case was not possible. Thus, three-point bending was used to assess tension-compression properties, while axial torsion was used to evaluate shear material properties.

Butterfly or tension wedge fractures are a typical fracture pattern in bending (Kress et al., 1995; Sharir et al., 2008; Turner, 2006) (Fig. 1), although other patterns, such as oblique and transverse fractures, have also been identified in experiments (Kress et al., 1995; Rich et al., 2005). The location of failure initiation in bending of the long bone is