Validation of a Cortical Bone Material Finite Element Model Under Three-Point Bending

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I. INTRODUCTION

The ability of human body models (HBMs) to accurately predict and represent cortical bone fracture is of paramount importance for the continued adoption of these models [1]. Historically, the material models used for cortical bone have generally not captured key aspects of the material behaviour [2], including material orthotropy, compression-tension asymmetry, and a complex failure mechanism. Recently, a material model (CFraC) has been developed by Cronin *et al*. [3] that accounts for many of these aspects of material behaviour to better model the loading and failure of cortical bone. The CFraC material model was validated for torsional and three-point bend loading using human femur test from the literature. The current study expands on this validation work by attempting to further validate the model under three-point bending using a novel test apparatus. The ultimate goal of this project is to model subject-specific tests using scanned geometry of each specimen. The work being presented here discusses the initial modeling, which was carried out to ensure that the modeling approach generally predicts the loading response and fracture pattern reported in the tests.

II. METHODS

Experimental Testing

A series of three-point bend tests on human femurs was carried out using a novel test apparatus (Fig. 1, left) which allowed rotation and axial translation of the potted ends of the femur diaphysis. After removing the epiphyses of the femurs, each end of the femur was set in potting that was 55 mm deep, with a span between the potting of 170 mm. Loading was applied to the centre of the femur span, and indenter force and displacement were measured along with rotation and displacement of the lower mounts of the supporting mechanism. A strain gauge mounted on the surface of the bone at the mid-span was used for comparison to the model results. The tension-side at the mid-span was at the anticipated location of failure initiation in the four experimental tests.

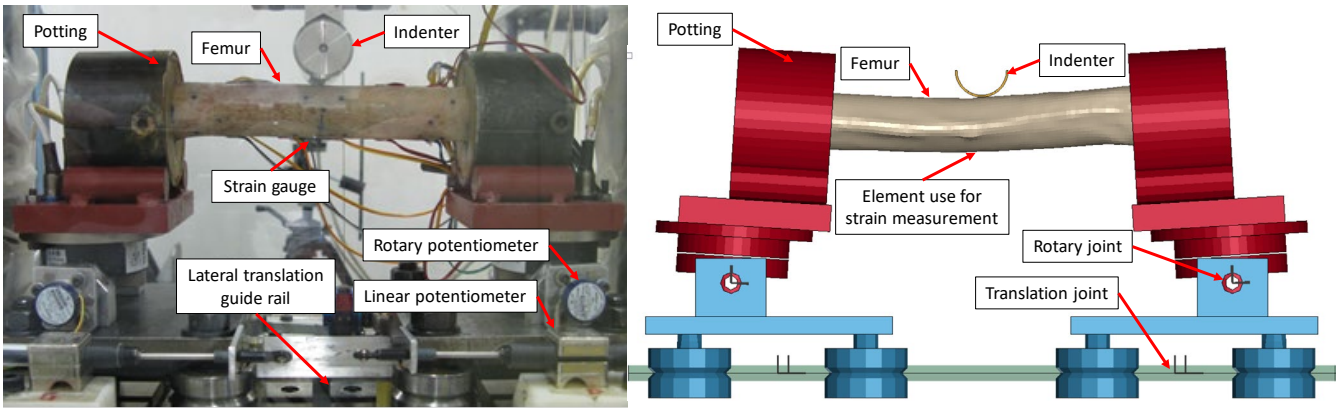

Fig. 1. Test setup prior to loading (left) and computation model immediately prior to failure (right).

Computational Modeling

A computational model (Fig. 1, right) of the test was developed and simulated using a commercial finite element solver (LS-DYNA R12.2, Ansys, Canonsburg, PA). Rigid bodies were used to model the 30 mm diameter indenter and support structure, with rotational and translational joints defined at the appropriate locations. The femur

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geometry was based on a CT scan carried out on one of the test specimens (Test P424), meshed using 2.5 mm hexahedral elements with three elements through the radial thickness of the bone and using the CFraC material model [3]. Nodes sets were defined on each end of the mesh, which were constrained to the steel rings surrounding the potting, to mimic the potting constraint in the model. A prescribed motion was applied to the indenter at a velocity matching the test (0.002 m/s), with contact defined between the indenter and bone to apply the load. As with the test, the indenter force and displacement, rotation and translation of the support structure, and longitudinal strain at the location of the strain gauge directly under the indenter were monitored.

III. INITIAL FINDINGS

A comparison of the model and test data (Fig. 2, left) showed good correspondence in the magnitude and shapes of the force, displacement, rotation and strain response to failure, although the initial slope of the strain-indenter displacement response was somewhat higher (51%) than the test response. The model prediction was 3.3% higher than the experimental failure value of the test indenter force (6.2 kN model vs. 6.0 kN test average), and less than the test indenter displacement (7.3 mm vs. 8.9 mm), support displacement (-6.0 mm vs. -6.8 mm), support rotation (3.6° vs. 4.3°), and strain at failure (2.1% vs. 2.2%), although all within the test variability.

In addition to the model predicting the quantitative responses of the test, the fracture pattern of the geometry being tested (Test P424) was well predicted by the model (Fig. 2, right), indicating that the mechanisms of fracture initiation and propagation were well captured by the material model.

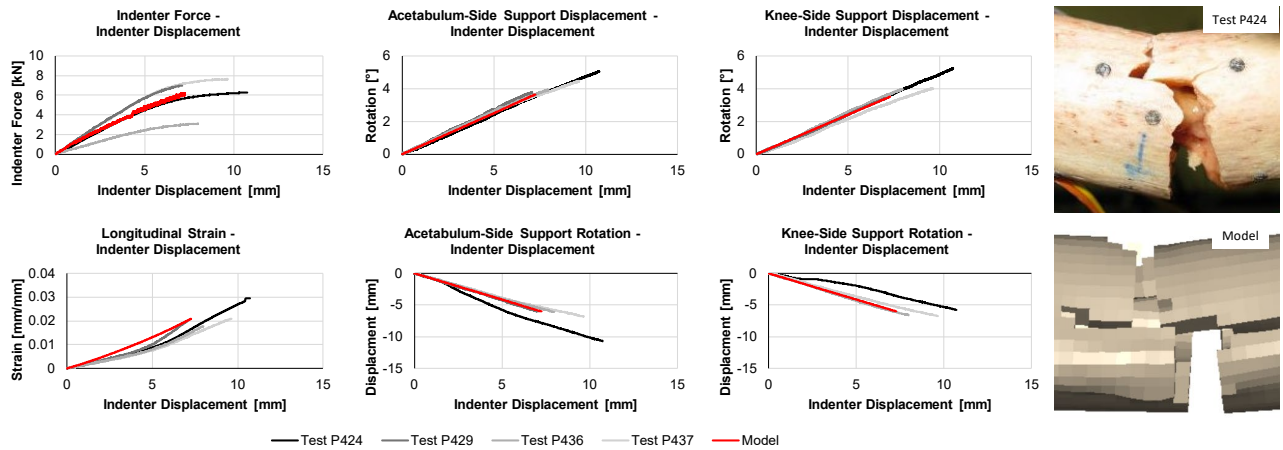

Fig. 2. Comparison of test and model kinematics (left), and Test P424 and model fracture pattern (right).

IV. DISCUSSION

The variability found in the test responses is generally within that which can be expected for biological materials. For all experimental tests, except Test P424, both the support rotation and translation responses were symmetric (left and right), indicating minimal sliding of the bone under the indenter. The strain measured during Test P424 was substantially higher (2.9%) than that of the average of the other three tests (2%), a finding that requires more investigation. Test P424 also exhibited asymmetric translation of the femur, indicating a 'settling' of the bone during test, possibly an effect of the larger displacement-to-failure.

The model matched the initial load up and initial plateau of the force-displacement response of Test P424 (the geometry chosen for the model) quite well, which indicates that the elastic portion and early damage phase of the material model are functioning well. The relatively long, flat portion of the force-displacement response of Test P424 was not captured particularly well by the model but this may be due to specimen-to-specimen variability in material properties of human femurs. Further investigation into the triaxiality (mean stress divided by effective stress) vs. effective strain at failure definition in the model and modeling other test-specific femurs is underway to quantify the responses for all tests.

V. REFERENCES

[1] Schmitt, *et al., Trauma Biomech*, 2019. [2] Khor, F., *et al., J Mech Behav Biomed Mater*, 2018. [3] Cronin, D., *et al.*, *Front Bioeng Biotechnol,* 2022.