COMPUTATIONAL HUMERUS MODEL: FRACTURE PREDICTION FOR ASSITIVE DEVICE USERS- LOFSTRAND CRUTCHES

Prateek Grover, MD, PhD candidate [1,2] , Mei Wang, PhD [1,2] ,
Peter Smith , MD [3] , and Gerald F. Harris, PhD, P.E. [1,2,3]

1. Medical College of Wisconsin, Milwaukee, WI 2. Marquette University, Milwaukee, WI and 3. Shriners Hospital for Children, Chicago, IL

INTRODUCTION

Finite element (FE) models of long bones have been used for evaluation of orthopaedic procedures such as trauma fixation[3] and arthroplasty[1]. Composite bones (Sawbones Worldwide, Pacific Research Labs, WA, USA) provide a reliable source of consistent geometry for FE models, as they allow for repeatable experimental validation of the FE model, and avoid handling and preservation issues associated with postmortem specimens. The current "fourth generation" material is proposed to mimic human cortical and cancellous bone properties.

Osteogenesis Imperfecta (OI), commonly called "brittle bone disease" is a heterogeneous group of connective tissue disorders, predominantly associated with a qualitative or quantitative Type I collagen defect, which primarily affect bone[4.] The global prevalence is approximately 1:10,000[6]. Major bone changes include low bone mass, disorganized micro-architecture and altered gross geometry, resulting in increased bone fragility. Other characteristic clinical features include blue sclera, deafness, short stature and multiple soft connective tissue manifestations [5].

OI bone fracture characteristics can be studied with FE models. The altered material properties of bone in OI can be combined with patient specific humeral geometry, loads and boundary conditions. Quantitative output information on load bearing ability of the humerus could then be used for activity modification, clinical care and therapeutic rehabilitation.

The goal of this study was to evaluate a long bone FE (Humerus) model. The specific aims included development of a Finite Element (FE) model of the fourth generation composite Humerus (HS4), followed by experimental evaluation and simulation for validation of the FE model. The model will be used to quantitatively characterize and predict fracture occurrence in subjects with OI, while using assistive devices.

METHODS

Design and development of the validated FE model included three phases:

1. FE Model Design

Volumetric humerus geometry, comprising mutually exclusive outer cortical and inner cancellous volumes, and a medullary cavity, was developed from transverse CT scans of a fourth generation composite Humerus, (HS4, Model 3404, Sawbones Worldwide, Pacific Research Labs Inc., WA, USA). Material properties of the composite bone, provided by the manufacturer, were assigned to the cortical and cancellous volumes. Local coordinate systems were defined based upon clinical Humeral Shaft Axis and Articular Margin Plane.

2. Experimental Evaluation

The FE local coordinate systems were adapted for accurately defining surfaces and planes for experimental evaluation. Three trials each of four-point bending were conducted for three specimens in two planes, AP (Antero-posterior) and ML (Medio-lateral), under a load range 0-400 N (equivalent bending moment 12.8 Nm). The compressive surfaces were the anterior surface in AP bending, and the medial surface in ML bending. Forces evaluated under displacement control were used to calculate construct stiffness and rigidity.

In addition, one specimen was instrumented with four stacked rosettes (C2A-06-062WW-350, Vishay Micro-Measurements, NC, USA) in the mid-diaphysial transverse plane on the anterior, posterior, medial and lateral surfaces. Strains acquired from these rosettes were used to calculate maximum and minimum principal strains.

3. FE Simulation

The 4-point upper roller loads and lower roller supports were incorporated in the FE model. Loads and boundary conditions were applied at geometric keypoints. The model was meshed with quadratic tetrahedral elements. A linear static analysis was performed, supported by experimental evidence. Convergence studies in strain and displacement were then completed.

RESULTS

1. Experiment Results

The force-displacement best fit was linear for all three specimens, with R^2 values greater than 0.999. The average stiffness and rigidity were greater in the ML plane than the AP plane. Interspecimen variability was less than 2% for both planes. A representative load displacement plot of one HS4 in both AP and ML 4-point bending is presented in Fig 1.

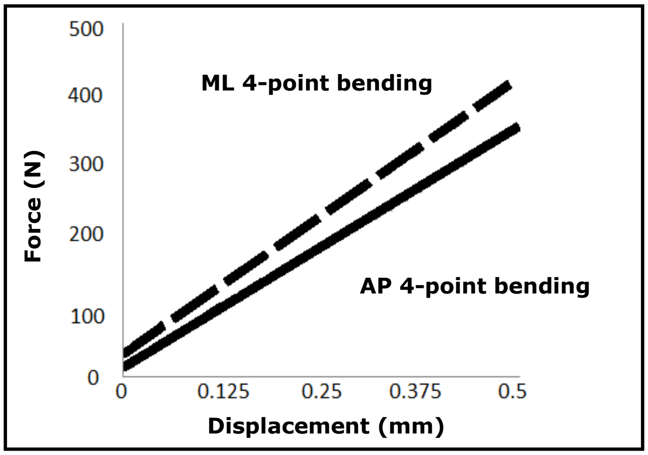

Fig 1. Representative Load-displacement plot of an HS4 in AP and ML 4-point bending

The maximum and minimum principal strains occurred at the experimentally defined tensile and compressive surfaces, respectively. A linear fit between strain and applied load was also noted.

2. Preliminary FE Simulation for AP bending

The HS4 deformed maximally in the mid-diaphysis, in the plane of loading. Principal strains were minimum along a plane normal to the loading (and support) plane, corresponding with the neutral axis. Maximum compression and tension occurred on the loading and support surfaces, respectively. The FE simulation results for AP bending are presented in Fig 2.The linearly increasing trend from the neutral axis to the surface, as well as the strain magnitudes correlated well with the uniaxial bending beam theory.

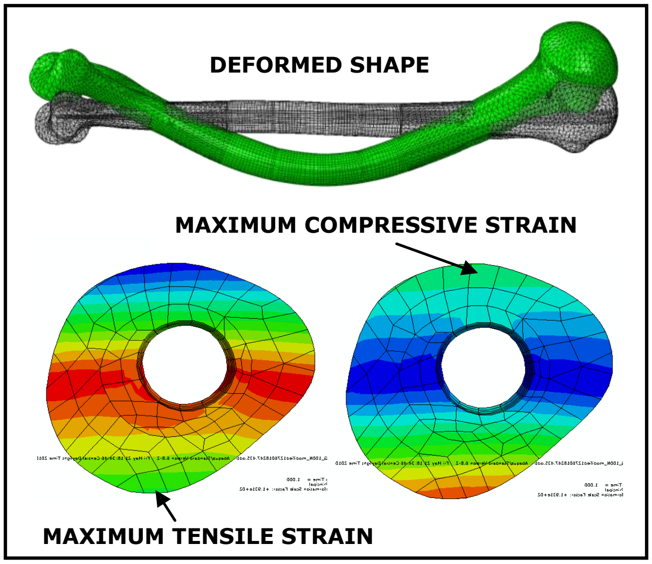

Fig 2. Preliminary FE results for AP bending (a) Deformed shape (b) Maximum principal strains

3. Comparison of Experimental and FE Simulation Results

Ongoing work involves further experimental validation and refinement of the FE model by assessing sensitivity to material properties (longitudinal modulus) for a range of 8–16 GPa. Experimental composite tests indicate a preliminary modulus value of 11 GPa. Simulations using a modulus of 8 GPa provide the best fit with experimental results.

No significant change in strains is noted by modeling the material as transversely isotropic.

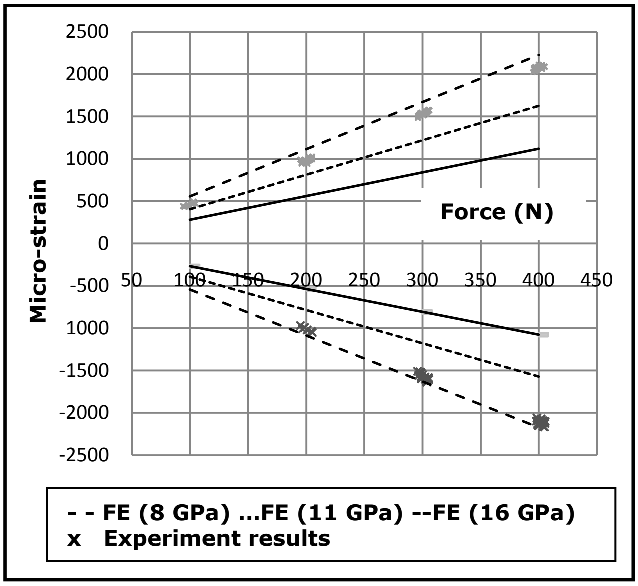

Fig. 3 Plot of FE (lines) and experimental (data points) principal strains vs. test load range (N)

DISCUSSION

The HS4 demonstrates linear behavior in AP and ML 4-point bending. R^2 values exceed 0.99 for both stiffness and strain. The model is stiffer in ML bending than in AP bending, which is similar in trend to other studies published in the literature[2]. The slightly greater ML diameter, leading to a greater moment of inertia about the neutral (bending) axis could explain the lower AP stiffness.

The model's linear experimental supported the linear FE analysis. Preliminary FE findings, including deformed shape and mid-diaphysial strains, agree in trend with the experimental model. Ongoing work involves assessing the sensitivity of the model to material properties, geometric factors, loading and boundary conditions. The sensitivity analysis points to maximum sensitivity to the longitudinal modulus among material properties, followed by cortical thickness among geometric properties.

The study provides mechanical characterization of fourth generation composite humerus behavior, which has not been published previously. The FE model is being developed to evaluate fracture risk in children with OI using assistive devices to ambulate. Other potential applications include evaluation of rehabilitative and orthopaedic procedures, and development of patient specific FE models.

CONCLUSION

Novel data on mechanical properties of the fourth generation composite humerus has been reported. This experimental data will be used to validate the FE model that has been developed. Best fit for simulations with experimental data is obtained at a longitudinal modulus of 8 GPa. Future applications of the validated FE model may include additional orthopedic pathology and rehabilitative management.

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REFERENCES

[1] A. Completo, F. Fonseca, J.A. Simoes, "Finite element and experimental cortex strains of intact and implanted tibia," Journal of Biomechanical Engineering vol. 129 (5), pp. 791-797, 2007

[2] J.T. Dunlap, A.C.M. Chong, G.L. Lucas, F.W. Cooke, "Structural properties of a novel design of composite analogue humeri models," Annals of Biomedical Engineering 36 (11), pp 1922-1926, 2008.

[3] M.P. Gardner, A.C.M. Chong, A.G. Pollock, P.H. Wooley, "Mechanical evaluation of large- size fourth-generation composite tibia and femur models," Annals of Biomedical Engineering, vol. 38 (3), pp. 613- 620, 2010.

[4] E. Martin E, J.R. Shapiro, "Osteogenesis Imperfecta: epidemiology and pathophysiology," Current Osteoporosis Rep, vol. 5(3), pp. 91-7, 2007

[5] D. Primorac, D.W.Rowe, M. Mottes, I. Barisić , D. Anticević , S. Mirandola et al., "Osteogenesis Imperfecta at the beginning of bone and joint decade," Croatian Medical Journal, vol. 42(4), pp. 393-415, 2001.

[6] P.J. Roughley, F. Rauch, F.H. Glorieux "Osteogenesis Imperfecta – clinical and molecular diversity," European Cells and Materials, vol.5, pp. 41 – 47, 2003.