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1-1-2013

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Jessica M. Fritz
Marquette University, jessica.fritz@marquette.edu

Nicole M. Grosland
University of Iowa

Peter A. Smith
Shriners Hospitals for Children

Gerald F. Harris
Marquette University, gerald.harris@marquette.edu

Published version. Published as part of *Proceedings of the American Society of Biomechanics,* 2013.

BRITTLE BONE FRACTURE RISK WITH TRANSVERSE ISOTROPY

[1,2]Jessica M. Fritz, [4]Nicole M. Grosland, [3]Peter A. Smith, and [1-3]Gerald F. Harris

[1]Marquette University, Milwaukee, WI, USA
[2]Medical College of Wisconsin, Milwaukee, WI, USA
[3]Shriners Hospitals for Children – Chicago, Chicago, IL, USA
[4]University of Iowa, Iowa, City, IA, USA
email: jessica.fritz@mu.edu

INTRODUCTION

Osteogenesis imperfecta (OI) is a heritable bone fragility disorder characterized by skeletal deformity and "brittle" bones. This fragility is believed to stem from a combination of bone mass deficiency and compromised bone material properties [1-3]. Poor bone quality poses major orthopaedic and rehabilitation challenges. Risk of fracture is a major consideration when prescribing activity restrictions and physical therapy. Quantifying bone fracture risk would be an invaluable clinical treatment tool. Finite element (FE) models have the potential to provide patient-specific feedback on the effects of fracture risk factors in long bones. As more information on OI becomes available, the model is evolving to include the latest data. The latest modification is focused on material properties. It is clear that OI bone does not behave like normal pediatric bone [1-5]. However, quantification of these differences has only recently been explored. Accurate material properties are essential for a computational fracture risk assessment (FE) model. The previous femoral fracture risk assessment model for an OI patient implemented material properties obtained during nanoindentation testing of OI bone specimens [5]. At the time, this was the only data available on OI bone properties. Based on these results, the femur was modeled as an isotropic material. Recent mechanical testing by our group has shed new light on the flexural properties of OI bone. These tests have shown that OI bone is, in fact, not isotropic and has a much lower Young's modulus (E) than what was calculated via nanoindentation testing. Transverse isotropic properties have been implemented into the OI femur model to examine effects on the maximum principal stress experienced during mid-stance of the gait cycle (highest load phase). The goal of study is to compare the isotropic and transverse isotropic models.

METHODS

A previously developed hexahedral FE model of an OI femur was used for this study [4]. The loading and boundary conditions assumed the mid-stance phase of gait; the condylar contact surfaces were fixed in all directions, while the femoral loads and muscle forces were derived from the kinetics of gait of a 12-year-old OI type I subject who underwent gait analysis at Shriners Hospital for Children – Chicago (Fig. 1).

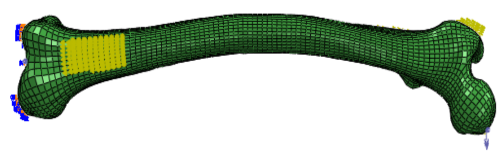

Figure 1: FE model of OI type I femur.

The material properties were assigned to reflect the new OI bone property data. The current FE femur is modeled with transversely isotropic material properties as shown in Table 1.

Table 1: Young's modulus (E, GPa), Poisson's ratio (υ) and shear modulus (G, GPa) for FE model of OI type I femur.

	$E_{11/22}$	E_{33}	$\upsilon_{11/22}$	υ_{33}	$G_{11/22}$	G_{33}
Cortical	4.0	7.0	0.3	0.3	1.5	2.7
Cancellous	3.0	6.0	0.3	0.3	1.2	2.3

The model was analyzed for three different levels of bowing: 5 mm, 15 mm and 25 mm. Maximum and minimum principal stresses were assessed and compared to previous isotropic results.

RESULTS AND DISCUSSION

The results of the maximum and minimum principal stresses and the percent difference between the isotropic and transverse isotropic models are shown in Table 2. As expected, the maximum and minimum principal stresses increased with the new model material properties. Interestingly, the percent difference of maximum principal stress was consistent for all levels of bowing around 10%; however, the minimum principal stress showed greater variation in percent difference as bowing increased. Percent difference of minimum principal stress ranged from 11.5% to nearly 24%.

Table 2: Results of principal stress comparison. An * depicts the transverse isotropic model.

	Maximum Principal Stress (MPa)	Minimum Principal Stress (MPa)
5mm	46.10	-43.50
5mm*	50.75	-48.50
% Diff	10.09	11.49
15mm	46.80	-45.00
15mm*	51.25	-50.60
% Diff	9.51	12.44
25mm	47.50	-47.10
25mm*	52.00	-58.40
% Diff	9.47	23.99

The principal stress results also show distinct differences between the two models. This can be seen in their stress contour plots (Fig. 2). The general location of the extreme stress values remains relatively unchanged between the isotropic and transversely isotropic models. However, the current model shows a very slight distal and anterior shift in the extreme stress areas as well as smaller areas of higher stress. The comparison is between a previous model with higher cortical and cancellous Young's modulus values of 19 GPa and 17 GPa, respectively. The effect of isotropy was examined in the 15 mm bowing model by comparison to the transversely isotropic model. The percent difference in maximum and minimum principal stresses showed a 5% and 7% increase, respectively, between the isotropic and transversely isotropic models.

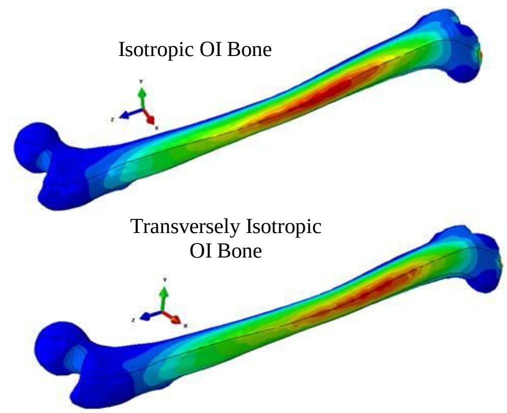

Figure 2: Contour plots of maximum principal stress on 15 mm bowed femur for previous (top) and current (bottom) FE models. Stress levels: red>blue.

CONCLUSIONS

Having a reliable fracture assessment model requires the most accurate input data available. Until recently, OI bone testing had been reported to be closer to an isotropic material than the anisotropic properties of normal bone. With new information indicating that this is not the case, the femur model was reconfigured and assessed to determine the effects of the newly acquired OI bone properties. As expected, the lower E values increased stress values. The femur being modeled as transversely isotropic rather than isotropic also affects the principal stresses due to loading during gait. Increased stresses lead to greater deviation towards fracture risk. This work updates the OI femur model to include recent biomaterial findings.

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ACKNOWLEDGEMENTS

Department of Education, NIDRR grant no. H133E100007. Contents do not necessarily represent the policy of the Department of Education.