12

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Nonlinear Viscoelastic Finite Element Modeling of a Female Pubic Symphysis

Z. Li, J-E. Kim, and A.W. Eberhardt

ABSTRACT

In this study, a three-dimensional nonlinear finite element (FE) model of a female pubic symphysis was developed using automatic mesh generation techniques for tetrahedral and hexahedral elements. The geometry was based on CT scan data of a female cadaver pelvis. The model was composed of cortical and trabecular bone, a fibrocartilaginous interpubic disc, and four ligaments connecting the pubic bones. Cortical and trabecular bone were assigned linear elastic properties, while properties for the soft tissues were estimated heuristically until the overall symphysis structural behaviors agreed with the results of previously published biomechanical experiments. The nonlinear hyperelastic James-Green-Simpson constitutive material equation was used to capture axial tension and compression, while a two-term Prony series was fit to the experimental creep data. The accuracy of the estimated material properties was further studied by comparing model predictions with experimental data from slow and fast tensile tests. The results demonstrated that the two-term Prony series extension of the James-Green-Simpson material model captured the nonlinear viscoelastic behavior of the pubic symphysis to within 6% of the experimental data. Element performance tests indicated that quadratic tetrahedrons and linear hexahedrons performed roughly the same in terms of solution accuracy and computation cost.

INTRODUCTION

The pubic symphysis connects the anterior portion of the human pelvic ring. It is a cartilaginous and slightly movable joint interposed with a fibrocartilaginous disc and supported by the superior, inferior, anterior, and posterior ligaments (Gamble et al., 1986). These ligaments are composed of fiber layers that blend with the interpubic disc (Clemente, 1985). Lesions of the pubic symphysis are common in victims of blunt trauma, such as motor vehicle collisions (Alonso et al., 1996; Weber et al., 1999). Clinically, pelvic pain caused by pubic symphysis laxity is a common symptom (LaBan et al., 1978). Previous work has demonstrated that lateral impact loading to the greater trochanter may lead to increased laxity of the pubic symphysis (Dakin et al., 2001).

Finite element (FE) models have been used to investigate pelvic stresses during normal ambulation (Dalstra and Huiskes, 1995) and during lateral loading (Renaudin et al., 1993; Plummer et al., 1998; Dawson et al., 1999; Majumder et al., 2004). Furthermore, models have been used to study the effects of different support conditions and loading rates on pelvic stresses in lateral impact simulations (Plummer et al., 1998; Majumder et al., 2004). Dawson et al. (1999) modeled impact of the pelvic bone structure for peak forces in the range of real world accidents. Recently, Anderson et al. (2003) developed a 3D FE model of a hemi-pelvis where cortical and trabecular bone were represented as linear elastic and isotropic, and acetabular cartilage was modeled as a two-parameter hyperelastic Mooney-Rivlin material. None of these computational models included a biofidelic pubic symphysis validated against experimental data.

The objective of this work was to create a realistic FE model of a female pubic symphysis and to validate the model using experimental data from Dakin et al., (2001). The James-Green-Simpson nonlinear constitutive material equation was used to predict the mechanical behaviors of the interpubic disc and pubic ligaments. Finite element material identification methods were used to estimate material properties of the soft tissues by simulating experimental force-displacement data (Flynn et al., 1998; Tönük and Silver-Thorn, 2003). Our intent is to ultimately incorporate the validated symphysis into a full pelvis model to investigate traumatic injuries of the pubic symphysis and pelvic bones under side impact loading conditions.

METHODS

Finite element model

CT data of a female human pubic symphysis (Figure 1(a)) was cropped from image data of the whole pelvis approximately 5 cm away from the midplane of the pubic symphysis. The cortical bone, underlying trabecular bone, and interpubic disc were segmented interactively, based on the voxels of CT image data. The anterior, posterior, inferior and superior ligaments were simplified as four separate solid bands based on the anatomical specimen geometry and manually added as a separate component into the segmented CT data before mesh generation (Figure 1(b)). These pubic ligaments were blended with the interpubic disc and outer surface of the cortical bone at insertion points. Triangular surfaces were first created using the marching cube algorithm (William and Harley, 1987). Then, these surfaces representing the segmented components were meshed into different volumetric tetrahedral elements using the advanced front method (Frey et al., 1996). To be in accordance with the experimental setup in Figure 2(a), the FE model was truncated to replicate the experimental conditions of axial tension, axial compression, and tension creep (Figure 2(b)). A total of 55,900 tetrahedral elements were included in this FE model.

To investigate concerns regarding the use of automatic tetrahedral meshing versus a more manually intensive hexahedral meshing in the finite element analysis, several FE models were investigated. Linear (4-node) and quadratic (10-node) tetrahedrons, as well as 8-node hexahedrons, were created and tested to investigate the influence of these element types on the accuracy of solutions. Currently, complex geometries can be discretized into tetrahedral elements more easily when compared to hexahedral elements (Cifuentes and Kalbag, 1992), which has increased the popularity of the automatic tetrahedral mesh generators. In the present study, the linear tetrahedral element meshes were created first using the automatic mesh generator in Amira 3.1 (TGS, Inc., San Diego, CA). The resultant meshes were then input into Hypermesh software (Altair Engineering, Inc., Troy, MI) for pre-processing to create the quadratic tetrahedral element model and export an output file for the ANSYS software (ANSYS, Canonsburg, PA). The ANSYS nonlinear structural analysis code was used to performed FE analyses of axial tension, axial compression, tensile creep, and loading rate analyses. The FE predicted displacement and total nodal reaction forces were output for comparisons with the experimental data (Dakin et al., 2001).

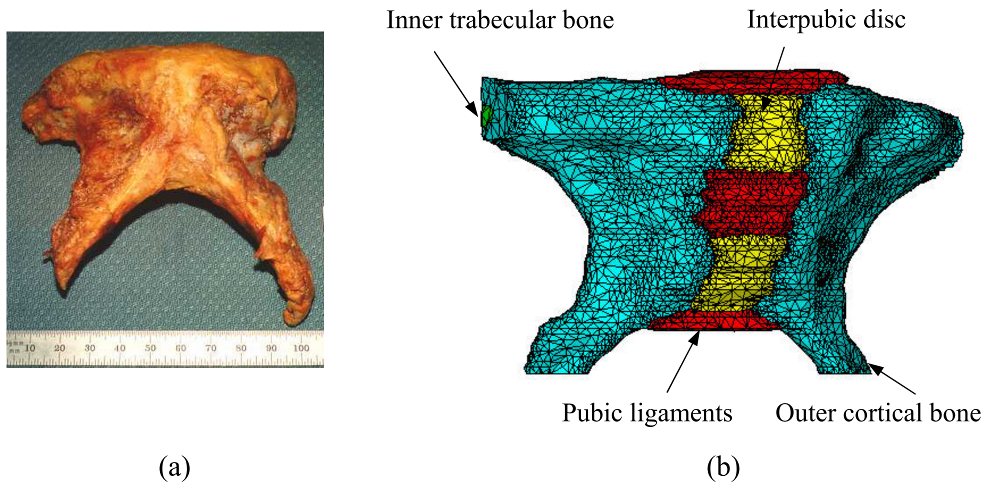

Figure 1: (a) A female pubic symphysis specimen (Dakin et al., 2001), (b) Volumetric tetrahedral finite element mesh of cortical bone, trabecular bone, interpubic fibrocartilaginous disc and four pubic ligaments.

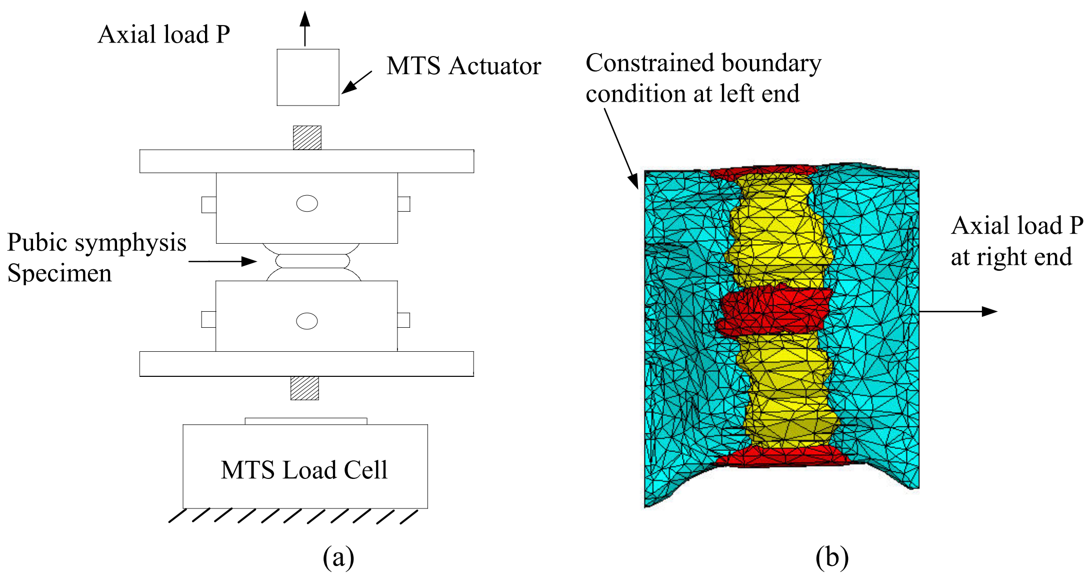

Figure 2: (a) Experimental set-up for tension/compression testing of the potted pubic symphysis joint (Dakin et al., 2001), (b) Quadratic tetrahedral finite element mesh of the truncated pubic symphysis consistent with experimental boundary conditions.

The hexahedral element models were then generated by splitting the tetrahedrons. However, this operation may result in bad element shapes for some hexahedrons. To solve this problem, Hypermesh software was further used to check and remove the hexahedral elements with bad shapes (very small and/or large internal angles). This is a relatively straightforward way to generate volumetric hexahedral elements from tetrahedral elements where segmented material identification numbers were kept for assigning material properties.

Initial estimation of Prony series constants

The heuristic search process for suitable material coefficients began with the use of a two-term Prony series, based on a generalized linear Kelvin-Voigt viscoelastic material model (Bradshaw et al., 1997; Tönük et al., 2004), to curve-fit the experimental creep data from Dakin et al. (2001), according to

$$d(t) = d_0\left[1 + a_1(1 - e^{-\frac{t}{\tau_1}}) + a_2(1 - e^{-\frac{t}{\tau_2}})\right] = \sum_{i=0}^{2} d_i \phi_i(t)\,. \tag{1}$$

Here d_0 is the instantaneous displacement prior to creep, τ_1, τ_2 are short-term and long-term creep time constants, and a_1, a_2 are short-term and long-term creep magnitudes. $\phi_i(t)$ is the basis function, such that

$$\phi_i(t) = (1 - e^{-\frac{t}{\tau_i}}) \qquad (\text{i} = 1, 2) \tag{2}$$

and $\phi_0(t) = 1$. The Prony coefficients that minimize the error can be acquired using a general linear least squares algorithm below (Eq. 3). The data to be fit, $d(t)$, is given as data pairs (t_p, d_p). The χ^2 error between a desired fitting function, $d(t)$, and the experimental data may be given by:

$$\chi^2 = \sum_{p=1}^{20}\left(\frac{d(t_p) - d_p}{d_p}\right)^2 \tag{3}$$

For a real viscoelastic material, all Prony coefficients should be positive so that physical and thermodynamic principles are satisfied. A sign control method was utilized, therefore, to ensure that all coefficients are positive (Bradshaw and Brinson, 1997). Using the experimental creep data (Dakin et al., 2001) where experimental value of d_0 was 0.64 mm, the initial values of Prony series coefficients (a_i, τ_i) were determined, which were: a_1 = 0.021, τ_1 = 2.24 sec., a_2 = 0.335, and τ_2 = 53.75 sec. These constants were then implemented in the FE model to estimate the nonlinear hyperelastic material coefficients of the interpubic disc and the ligaments.

Constitutive models and material constant estimation for nonlinear hyperelastic behavior

The exact anisotropic material properties for the fibrocartilaginous disc and pubic ligaments, however, have not been well described. Presently, the James-Green-Simpson nonlinear hyperelastic material constitutive model was used to capture the nonlinear elastic response of the soft tissues with an extension to simulate viscoelastic behavior (Simo, 1987; Tönük and Silver-Thorn, 2004). This model is a modified 9-parameter Mooney-Rivlin material model. The time-dependent strain energy function (energy per unit volume), W(t), with two-term Prony constants in creep form, can be expressed by:

$$W(t) = W_0[1 + a_1(1 - e^{-\frac{t}{\tau_1}}) + a_2(1 - e^{-\frac{t}{\tau_2}})] \tag{4}$$

W_0 is the instantaneous strain energy function (before creep) for an incompressible hyperelastic material, represented by

$$W_0 = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) + C_{11}(I_1 - 3)(I_2 - 3) + C_{20}(I_1 - 3)^2 + C_{30}(I_1 - 3)^3 \tag{5}$$

where I_1 and I_2 are the first and second invariants of the Green-Lagrange strain tensor. C_{ij} (i=0, 3; j=0, 1) is the material constant to be determined based on experimental data. With a further assumption of axisymmetry: $I_1 = I_2 = I$, $C_{10} = C_{01} = 0.5C_1$, $C_{11} = C_{20} = 0.5C_2$, and $C_{30} = C_3$. This strain energy function can be simplified to:

$$W_0 = C_1(I-3) + C_2(I-3)^2 + C_3(I-3)^3 \tag{6}$$

C_1 represents the initial stiffness of the soft tissues at low strain, while C_2 accounts for the stiffening of the tissues with increasing strain, and C_3 causes the increased stiffening of the tissues at large strain (Tönük and Silver-Thorn, 2003). This reduced constitutive model was used to model nonlinear viscoelastic material properties of the interpubic fibrocartilaginous disc and ligaments. The resulting coefficients were the time-independent material constants C_i (i = 1, 2, 3), and time-dependent Prony series constants a_i , τ_i (i=1, 2).

Ligaments consist largely of parallel-bundled collagen fibers embedded in a ground substance matrix holding large amount of trapped water. We assumed that the ground substance dominated the compressive load response of the pubic ligaments. To simplify the search process for material estimation, the ground substance was assumed to have the same material property (C_1=1.44 MPa) as that of the medial collateral ligament (MCL) of the knee joint (Gardiner and Weiss, 2003). This constant was not modified during subsequent material constant estimation for the interpubic disc and the ligaments using FE methods.

The material coefficients for the interpubic disc were estimated from experimental force-displacement data of compression tests (Dakin et al., 2001) on female pubic symphyses. Initial trial values of constants C_1, C_2 and C_3 were based on values of Young's modulus and aggregate modulus of articular cartilage, as reported in the literature (Athanasiou et al., 1991; Wu et al., 1998; Herzog et al., 1998; Li et al., 2000; Nieminen et al., 2004). These constants were then manually modified during an iterative search process using the FE model, until the normalized least square errors comparing the experimental data and FE calculated data were within 6%, according to Tönük and Silver-Thorn (2004), where

$$\chi^2 = \sum_{i=1}^{20} \left(\frac{F_{fe_i} - F_{\exp_i}}{F_{\exp_{\max}}} \right)^2 \tag{7}$$

Here F_{fe_i} is the finite-element reaction force at time step i, $F_{\exp_i}$ is the experimental reaction force at time step i, and $F_{\exp_{\max}}$ is the maximum reaction force during tension or compression tests. The Prony constants and ligament material coefficients were not modified during the iterative process.

The remaining material constants (C_2 , C_3) for the pubic ligaments were then estimated using experimental data of force-versus-displacement in tension (Dakin et al., 2001). The procedures were repeated to search for ligament properties while the Prony constants and cartilage disc properties were not altered. During material constant estimation by FE methods, cortical and trabecular bones were modeled as linear isotropic elastic materials. Young's modulus and Poisson ratio were directly selected from the literature as listed in Table 1.

Table 1. Isotropic Linear Elastic Material Constants For Cortical And Trabecular Bones.

Material	**Young's Modulus (MPa)**	**Poisson Ratio**	**References**
Cortical bone	17000	0.3	Dalstra and Huiskes,1995
Trabecular bone	100	0.2	Dalstra et al., 1993

Creep analysis

Using the estimated material constants for the soft tissues and the initial Prony constants, tensile creep was again simulated iteratively to refine the Prony constants. Loading and boundary conditions were applied to simulate the experiments done by Dakin et al., (2001), where the bony ends of pubic symphysis specimens were potted in acrylic cement so that the interpubic disc was perpendicular to the loading axis. Mechanical tensile creep tests in that study included a load-and-hold test to record creep displacement of the pubic symphysis for 60 seconds. To agree with the creep experimental set-up, the six degrees of freedom of the nodes on the left end of FE model (Figure 2(b)) were constrained, and axial nodal forces was applied to the nodes on the right end to achieve the instantaneous displacement of 0.64 mm, and then held for 60 seconds. Using the estimated material properties, experimental tensile creep data, and the initial Prony constants, FE creep analysis was performed to get the new Prony constants a_1' and a_2' by manually modifying the initial constants a_1 and a_2, while the time constants τ_1 and τ_2 were not changed during this search process. The final values of the Prony constants were determined when the error was less than 1%.

Loading rate analysis

Finally, using the estimated material constants and the new Prony series coefficients, the effects of loading rate were examined by simulating two tensile test conditions (0.01 mm/s and 100 mm/s loading rates) using the FE model (Figure 2(b)). The boundary conditions of tensile loading rate analysis were the same as that of tensile creep analysis. The nodal forces were ramp-loaded to reach the axial displacement of 2.5 mm at different rates by controlling the time at the end of load step and number of load substeps during FE analysis. This served as a test to the FE model to verify accuracy of the estimated material constants.

RESULTS

The creep experiments were best simulated using the resulting two-term Prony constants (a_1' = 0.074, τ_1 = 2.24 sec., and a_2' = 0.306, τ_2 = 53.75 sec.) and the estimated material coefficients in Table 2. The FE-predicted creep data were consistent with experimental data with normalized error less than 1%, as shown in Figure 3. The overall load-displacement curves from FE predicted and experiments for ramp loading and unloading process were in good agreement with the experimental data (total normalized errors less than 6%), as plotted in Figure 4.

Table 2. Estimated Hyperelastic Material Constants For Interpubic Disc And Pubic Ligaments.

Material	**C_1 (MPa)**	**C_2 (MPa)**	**C_3 (MPa)**
Pubic disc	0.60	1.1	0.25
Pubic ligaments	1.44	0.57	50.0

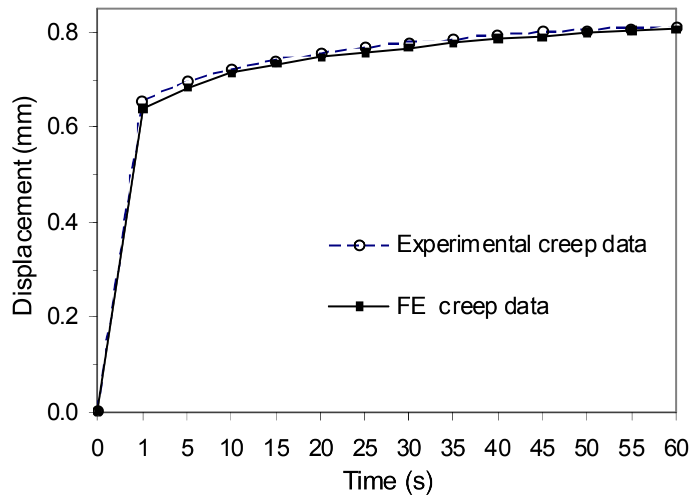

Figure 3: Displacement-time curve obtained from experimental creep test and FE creep analysis. The total normalized error between FE data and experimental data was within 1%.

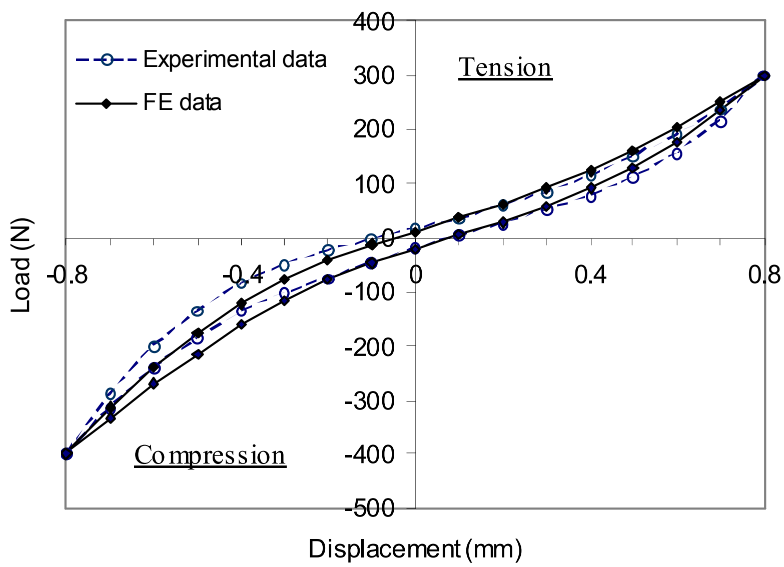

Figure 4: Load-displacement curve obtained during tension-compression experiments and FE analysis by ramp-loading the joint at rate of 1mm/s to low amplitude of ±0.8 mm. The total normalized error between FE data and experimental data was within 6%.

The finite element results for the two different loading rates (0.01 mm/s and 100 mm/s) indicated that the pubic joint exhibited rate-dependent behaviors (Figure 5). The predicted load-displacement curves revealed peak tensile load of 1.22 kN for loading rate 0.01 mm/s and 1.52 kN at 100 mm/s, for the average displacement of 2.5 mm, which agree well with the experimental values found by Dakin et al. (2001).

The results demonstrated that the finite element model predictions of the quadratic tetrahedron and linear hexahedron meshes were roughly equivalent in terms of accuracy and computation cost, as concluded by Cifuentes and Kalbag (1992). For the present pubic symphysis model, solutions from the highly refined 4-node tetrahedral element model (total 55900 elements) were acceptable (within 8%) of solutions from the 10-node tetrahedral model (total 20223 quadratic elements). Thus, the quadratic tetrahedral element model was selected for the present models, for which it was difficult to generate using hexahedral elements.

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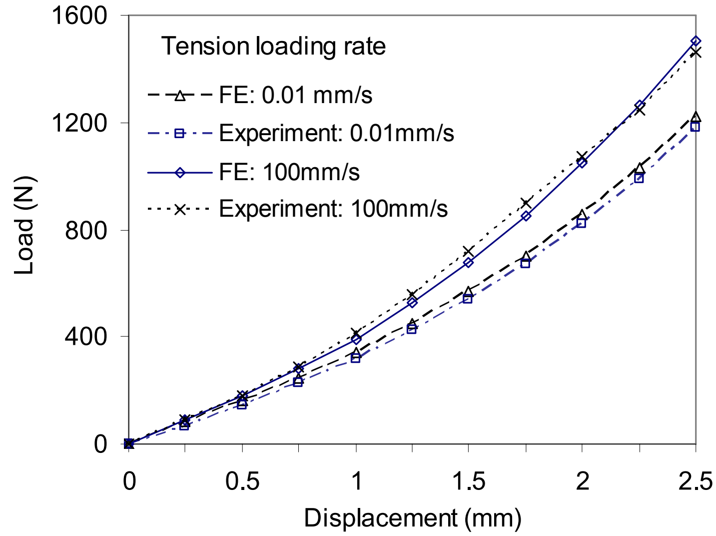

Figure 5: Reaction force-displacement curve obtained by FE analysis and experiments at the loading rates of 0.01 mm/s and 100 mm/s. The total normalized error between FE data and experimental data was less than 2%.

DISCUSSION

This paper demonstrated an effective method to construct a complicated three-dimensional biofidelic FE model with validated nonlinear visco-hyperelastic material properties. The viscoelastic extension of James-Green-Simpson nonlinear material models approximated quasistatic tension and compression, tensile creep, and the loading rate-dependent behavior of the pubic symphysis within 6% of experimental data to our satisfaction. The biological tissues that comprise the pubic symphysis are complex both in composition and load response; therefore, a few assumptions were made associated with the computational results in this study. These included (i) identical material properties in tension and compression for the cartilage disc; (ii) the nonlinear constitutive model for the ligaments was unable to consider the fiber stretching direction; (iii) axisymmetry assumption in material model is not real for actual soft tissues and may compromise the accuracy of the estimated material constants.

The material parameters (C_1, C_2, C_3) and the Prony series constants (a_1, a_2, τ_1, τ_2) of the soft tissues estimated in this study were extrapolated for one individual geometry of a female pubic symphysis; therefore, they should not necessarily be considered as material constants to be input in other FE models of the pubic symphysis. The experimental study done by Dakin et al. (2001) found that the overall structural mechanical response of the pubic symphysis joint varied with gender, age and individual geometry. We are confident, however, that the methods presented here provide some guidelines for others intending to quantify the mechanical behaviors of soft tissues and joints using experimental data.

In conclusion, the FE model of pubic symphysis achieved the intended goals. The FE computational results reasonably captured the average load response of a female pubic symphysis observed in previous experiments. We have incorporated the symphysis model into a FE model of a whole female pelvis, to continue ongoing investigations of traumatic injuries to the human pelvis resulting from automotive side impacts.

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REFERENCES

ALONSO, J. E., LEE, J., BURGESS, A. R., and BROWNER, B. D. (1996). The management of complex orthopaedic injuries. Surgical Clinics of North America, 76, 879-903.

ANDERSON, A. E., PETERS, C. L., TUTTLE, B. D., and WEISS, J. A. (2003). Development and validation of a finite element model of the pelvis. Proceedings of IMECE'03 2003 ASME International Mechanical Engineering Congress, Washington, D.C., November 15-21.

ATHANASIOU, K. A., ROSENWASSER, M. P., and BUCKWALTER J. A. (1991). Interspecies comparisons of in situ intrinsic mechanical properties of distal femoral cartilage. J. Orthop. Res., 9, 330-340.

BRADSHAW, R. D. and BRINSON, L. C. (1997). A sign control method for fitting and interconverting material functions for linearly viscoelastic solids. Mech. Time Dep. MATLS. 1, 85-108.

CIFUENTES, A. O. and KALBAG, A. (1992). A performance study of tetrahedral and hexahedral elements in 3-D finite element structure analysis. Finite Elements in Analysis and Design, 12, 313-318.

CLEMENTE, C. D. (1985). Pelvis, in: *Gray's anatomy*, Thirtieth American Edition, pp. 362-363.

DAKIN, G. J., ARBELAEZ, R. A., MOLZ, F. J., ALONSO, J. E., MANN, K. A., and EBERHARDT, A. W. (2001). Elastic and viscoelastic properties of the human pubic symphysis joint: effects of lateral impact loading. Journal of Biomechanical Engineering, 123, 218-226.

DALSTRA, M. and HUISKES, R. (1995). Load transfer across the pelvic bone. J Biomech., 28(6), 715-724.

DALSTRA, M., HUISKES, R., ODGAART, A., and VAN ERINING L. (1993). Mechanical and textural properties of pelvic trabecular bone. Journal of Biomechanics 26(4-5), 523-535.

DAWSON, J. M., KHMELNIKER, B. V., and MCANDREW, M. P. (1999). Analysis of the structural behavior of the pelvis during lateral impact using the finite element method, Accident Analysis and Prevention, 31, 109-119.

FLYNN, D. M., PEURA, G. D., GRIGG, P., and HOFFMAN, A. H. (1998). A finite element based method to determine the properties of plantar soft tissues. J. Biomed. Eng., 120, 202-210.

FREY, P. J., BOROUCHAKI, M., and GEORGE, P. (1996). Delaunay tetrahedralization using advanced-front approach. Proceedings of 5th international Meshing Roundataable, SAND'96, pp. 31-46.

GAMBLE, J. G., SIMMONS, S. C., and FREEDMAN, M. (1986). The symphysis pubis: anatomical and pathological considerations. Clin. Orthop. Relat. Res., 203, 251-272.

GARDINER, J. C. and WEISS, J. A. (2003). Subject –specific finite element analysis of the human medial collateral ligament during valgus knee loading. Journal of Orthopedic Research, 21, 1098-1106.

HERZOG, W., DIET, S., SUTER, E., MAYZUS, P., LEONARD, T. R., MÜLLER, C., WU, J. Z., and EPSTEIN, M. (1998). Material and functional properties of articular cartilage and patellofemoral mechanics in an experimental model of osteoarthritis. Journal of Biomechanics, 31, 1137-1145.

LABAN, M. M., MEERSCHAERT, J. R., TARLOR, R. S., and TABOR, H. D. (1978). Symphyseal and sacroiliac joint pain associated with pubic symphysis instability. Arch. Phys. Med. Rehabil., 59, 470-472.

LI, L. P., BUCSCHMANN, M. D., and SHIRAZI-ADL, A. (2000). A fibril reinforced nonhomogeneous poroelastic model for articular cartilage: inhomogenous response in unconfined compression. Journal of Biomechanics, 33,1533-1541.

MAJUMDER, S., ROYCHOOWDHURY, A., and PAL S. (2004). Dynamic response of the pelvis under side impact load – a three-dimensional finite element approach. IJ Crash, 9(1), 89-103.

NIEMINEN, M. T., TÖYRÄS, J., LAASANEN, M. S., SILVENNOINEN, J., HELMINEN, H. J., and JURVELIN, J.S. (2004). Prediction of biomechanical properties of articular cartilage with quantitative magnetic resonance imaging. Journal of Biomechanics, 37, 321-328.

PLUMMER, J. W., EBERHARDT, A. W., ALONSO, J., and MANN, K. A. (1998). Parametric tests of the human pelvis: the influence of load rate and boundary condition on peak stress location during simulated side impact. Advances in Bioengineering, ASME1998 BED-Vol. 39, 165-166.

RENAUDIN, F., GUILLEMOT, H., LAYASTE, F., SKALLI, W., LESAGE, F., and PECHEUX, C. (1993). A 3D Finite element model of pelvis in side impact, Proceedings of the 37th Stapp Car Crash Conference, SAE 933130, pp. 249-259.

SIMO, J. C. (1987). On a fully three-demensional viscoelastic damage model: formulation and computational aspects. Comput. Methods Appl. Mech. Eng., 1, 79-85

TÖNÜK, E. and SILVER-THORN, M. B. (2004). Nonlinear viscoelastic material property estimation of lower extremity residual limb tissues. Journal of Biomech. Eng., 126, 289-300.

TÖNÜK, E. and SILVER-THORN, M. B. (2003). Nonlinear elastic material property estimation of lower extremity residual limb tissues. IEE Trans. Neural Sys. Rehab. Eng., 11(1), 43-53.

WEBER, K., VOCK, B., MÜLER, W., and WENTZENSEN, A. (1999). Rupture of the pubic symphysis: diagnosis, treatment, and clinical outcome. Annals of Saudi Medicine, 19(6), 544-546.

WEISS, J. A. and GARDINER, J. C. (2001). Computational modeling of ligament mechanics. Critical Reviews™ in Biomedical Engineering, 29(4) 1-70.

WILLIAM, E. L. and HARLEY, E. (1987). Marching cubes: a high resolution 3D surface construction algorithm. Computer Graphics, 21(4).

WU, J. Z., HERZOG, W., and EPSTEIN, M. (1998). Evaluation of the finite element software ABAQUS for biomechanical modeling of biphasic tissues. Journal of Biomech., 31, 165-169.

DISCUSSION

PAPER: **Nonlinear Viscoelastic Finite Element Modeling of a Female Pubic Symphysis**

PRESENTER: ***Zouping Li, Department of Biomedical Engineering and the Department of Mechanical Engineering, University of Alabama at Birmingham***

QUESTION: *Erik Takhounts, NHTSA*

I actually don't remember whether NHTSA actually paid for that research, but anyways that's a question to somebody else. To you, I have a reservation. First of all, whatever you call a nonlinear viscoelastic material, what you presented is not because your stress/strain relationship is nonlinear, which you expressed in terms of strain density function. And then, you have a linear proni-series, which is a time-dependent function. So how do you add them up together? Do you convolute them? Do you just add up the two responses?

ANSWER: Yeah. You cut the material…overload the response for the pubic response region, nonlinear. So first, we adjust the linear viscoelastic model...some initial constant for …service. We're looking for some material constant where many may need to be changed until the FE results can match either the pyramid results.

Q: At this time, I have a question on that one, too, but [the] first one was a comment. You have to go back and look what actually can be called as nonlinear viscoelastic material and what cannot. The second question is with regard to your test methods. It is known that numerical methods, especially the tetrahedral elements that you used, which are very well known to be stiffer than the actual material response. Then, you use a finite element model, which is a not bad technique, to estimate your material constants. So, how do you incorporate whatever the numerical junk you get with your model into your material constants? Because if you use these material constants with some other model, they're going to change.

A: I agree with you because I just want…the methodology to estimate the material constant for the average property. For example, for each study, the premier cadaver pubic symphysis, you are…progress for the lotto…You get pyramid for female and the male pubic symphysis, which just average the pyramid and use reasonable… to estimate the property. Actually for different pair, pubic symphysis, the property may be changed. Right.

Q: Have you tried to do some kind of convergence study of your elements, the element definition study?

A: I would do this in the future because a lot of percentage. It's a true model of uncertainty; you use this methodology to view.

Q: This should be done before you present anything.

A: Yeah.

Q: Actually, in finding elements stress.

A: Yeah.

Q: Okay. Thanks.

QUESTION: *Guy Nusholtz, DaimlerChrysler*

The specimens that you're using have a complex, internal geometry. How do you know that the material response that you're getting is not a function of the geometry as opposed to the material properties? And if that is the case, then have you checked to see whether your material response is sample-dependent, size of the sample dependent?

A: Actually, we put this model from CT data and in the end, according to pubic symphysis, and according to CT data, I build this model. This process is very complicated. The CT data: You already have 200 or 300 slices for one whole pelvis. Firstly, before you could sort the model, you must recognize which area is the soft tissue or which area is the cortical or the trabecular bone. Actually from either ridge, you can find the area of the cortical bone, trabecular bone or soft tissues. So, you have side impact study, we use base value to try to make our whole pelvis model use, the weight of a whole pelvis model close to the model I generated

Q: Those are really…a material property is not an artifact of the method that you used to get the material property.

A: Yeah. I read the sample to reach above property of the cartilage disk and the ligament, so that's the reason I tried to use the high-watt, nonlinear material model to use here and the material property of pubic ligament and disk.

Q: Okay. Done.

QUESTION: *Andre Loyd, Duke University*

I have a question: Why did you use tetrahedral elements instead of something like hex, hexahedral elements? And, how many elements were in your model?

A: You mean for the pubic symphysis or the whole pelvis model?

Q: Yeah. Just the pubic symphysis.

A: For the—95.

Q: Okay. And the element types?

A: Tetrahedral elements.

Q: Yeah. Why did you use tetrahedral--?

A: To go contrary, I use the soft…

Q: Okay. Thank you.