# DEVELOPMENT AND VALIDATION OF A SUBJECT-SPECIFIC FINITE ELEMENT MODEL OF THE PELVIS: ASSESSMENT OF MODEL SENSITVITY

**A.E. Anderson[1], C.L. Peters[2], B.D. Tuttle[3], and J.A. Weiss[4,5]**

ABSTRACT

The objectives of this study were to: 1) develop and validate a realistic FE model of the pelvis using subject-specific bone geometry, location-dependent cortical thickness and trabecular bone elastic modulus, and 2) assess the sensitivity of FE predictions to assumptions in both assumed and measured model inputs. A finite element model of a cadaveric pelvis was created using subject-specific CT image data. The pelvis was loaded experimentally using a prosthetic femoral stem in a fashion that could be easily duplicated in the computational model. Cortical bone strains were monitored using rosette strain gauges in ten locations throughout the left hemi-pelvis. FE predictions were compared directly with experimental results for purposes of validation. Overall, FE predictions were consistent with experimental results. In addition, the results of the sensitivity study suggest that changes to cortical bone thickness had the largest effect on cortical bone strains. Deviations to all other material parameters had little effect on cortical strains.

INTRODUCTION

The pelvic joint is one of the most important weight bearing structures in the human body. Despite its efficient structure, the pelvis has little tolerance for abnormal loads [1]. Although the major problems associated with the pelvis are largely attributed to altered biomechanics, there is very little quantitative information concerning the in-vivo stress distribution in the pelvis. An in vivo approach to the study of pelvic stresses and strains may provide valuable information that could lead to improved implant designs, surgical approaches, diagnosis, and may provide the framework necessary for preoperative surgical planning.

The FE method has often been utilized to investigate various aspects of orthopedic medicine. The FE method is preferred over experimental analyses when it is necessary to model clinical abnormalities. This becomes clear when one considers how difficult (if not impossible) it would be to assemble a population of cadaveric tissue that exhibits a specific pathology. Nevertheless, the process to create subject-specific models remains questionable due to the fact that direct validation has not been performed. Furthermore, a complete sensitivity analysis has not been done to understand the mechanical significance of both assumed and measured inputs.

Keywords: pelvis, finite element model, biomechanics, strain gauges

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[1]Ph.D. Student, [3]Research Assistant, [4]Corresponding Author, Associate Professor, Department of Bioengineering, University. of Utah, 50 S. Central Campus Dr., Rm 2460 Salt Lake City, Utah, USA
[2]Associate Professor, [5]Adjunct Associate Professor, Department of Orthopedics, University. of Utah, 3B165 SOM, Salt Lake City, Utah, USA

## METHODS

A combined experimental and computational protocol was used to develop and validate a subject-specific three-dimensional model of a 68 y/o female pelvis.

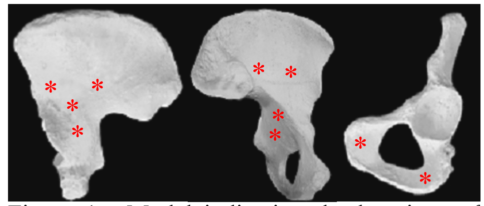

Figure 1: Model indicating the locations of the ten rosette strain gauges that were attached to the left cadaveric hemi-pelvis.

*Experimental Study.* The sacroiliac joint and all soft tissues, with the exception of articular cartilage, were removed. A registration block and wires were attached to the iliac crest. The block allowed for spatial registration of experimental and FE coordinate systems, while the wires served as a guide to reproduce the boundary conditions used in the experimental model [2]. A volumetric CT scan (512x512 acquisition matrix, FOV=225 mm, 354 slices, slice thickness=0.6 mm) was obtained in a superior to inferior fashion.

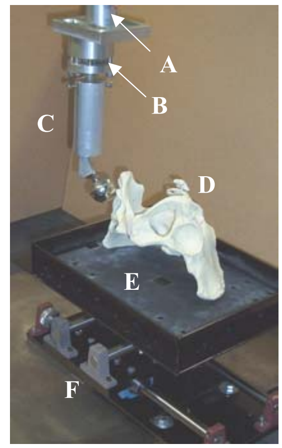


Figure 2: Fixture for loading pelvis via femoral component. A) load cell, B) ball joint, C) femoral component, D) pelvis, E) mounting pan for embedding pelvis, and F) X-Y table.

The mounting and loading of the pelvis followed the protocol of Dalstra et al. [3]. The iliac crests were submerged in a mounting pan of catalyzed polymer resin (Bondo, Mar-Hyde, Atlanta, GA) to the depth defined by the iliac guide wires. Ten three-element rectangular rosette strain gauges (WA-060WR-120, Vishay Measurements Group, Raleigh, NC), representing 30 channels of data, were attached to the left hemi-pelvis at locations around the acetabulum, pubis, ischium, and ilium (Fig. 1). Vertically oriented loads (0.25, 0.50, 0.75, and 1.0 BW) were applied to the acetabulum via a femoral prosthesis attached to a linear actuator, while strains were recorded continuously (Fig. 2). 3D coordinates of the strain gauges and registration block were determined with respect to the laboratory reference frame using an electromagnetic digitizer (Immersion Corp., San Jose, CA).

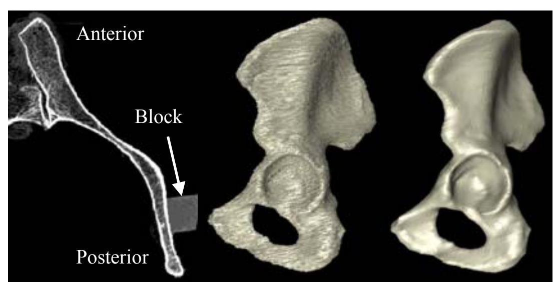


Figure 3: CT image slice at the ilium showing the radio-opaque registration block and the distinct boundary between cortical and trabecular bone (left). The original cortical bone surface was reconstructed by stacking the segmented contours (middle). This surface was smoothed and decimated to create the final surface (right).

*Geometry Extraction, Mesh Generation.* Contours for the outer cortex and the boundary of the cortical and trabecular bone, registration block, and guide wire were extracted from the CT data for the left hemi-pelvis via manual segmentation (Fig. 3). The points comprising the contours were triangulated to form the original surface, which was then decimated and smoothed to form the final surface using the Visualization Toolkit (VTK). The triangles composing the final surface were used to define a FE mesh of shell elements to represent cortical bone. A volumetric tetrahedral mesh was created using the FE surface mesh and the mesh generation program CUBIT (Sandia National Labs, Albuquerque, NM). The final FE model consisted of 210,000 tetrahedral solid elements for trabecular bone and 30,000 triangular shell elements for cortical bone (Fig. 4). Acetabular cartilage was represented with 500 shell elements with a constant thickness of 2 mm, determined by averaging the distance between the implant and acetabulum in the neutral kinematic position.

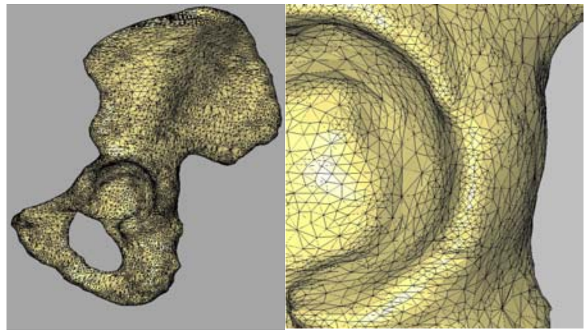

Figure 4: FE mesh of the pelvis (left) with a close-up view of the mesh at the acetabulum (right).

*Cortical Thickness.* An algorithm was developed to determine the spatially varying thickness of the cortex based on the distances between the polygonal surfaces representing the outer cortex and the boundary between the cortical and trabecular bone. Vectors were constructed between each node on the cortical surface and the 100 nearest nodes on the surface defining the cortical-trabecular boundary (Fig 5, left). Cortical thickness was determined by minimizing both the distance between the nodes of the surfaces and the angle of the dot product between the surface normal of the cortical surface with that of each corresponding trabecular vector. In areas of high curvature (such as the acetabular rim), special consideration of thickness was necessary (Fig. 5, right). When the above-described algorithm reported a thickness value that exceeded 1.5 times the smallest distance between the nodes, the smallest distance between nodes on the two surfaces was used to determine the cortical thickness. The minimum value of nodal thickness was assumed to be 0.44 mm or the width of one pixel (FOV= 225, FOV/512 = 0.44 mm/pixel). The algorithm was tested for accuracy using polygonal surfaces representing parallel planes, concentric spheres, and layered boxes with varying mesh densities.

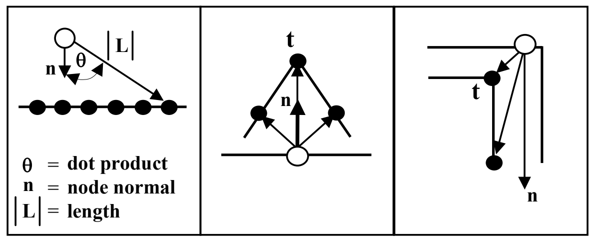


Figure 5: Schematics illustrating the cortical thickness algorithm with cortical nodes (open circles) and nearest trabecular nodes (filled circles). Left - the lengths of the nearest trabecular nodes and dot product were analyzed. Middle - the smallest angle of the dot product between the cortical node and nearest trabecular node neighbor yields the desired thickness measurement (**t**). Right - the normal vector from the cortical node does not intersect the trabecular surface. In this situation a weighting scheme was applied such that the smallest distance between the nodes was taken as the cortical thickness.

*Material Properties, Boundary Conditions, & Analysis.* The femoral implant was represented as rigid while cortical and trabecular bone were represented as isotropic hypoelastic. Initial material properties for cortical bone were $E$=17 GPa and $\nu$=0.29 [3]. Relationships between CT Hounsfield unit, apparent density and elastic modulus were used to assign a density-dependent modulus for each tetrahedral element [4,5]. Acetabular articular cartilage was represented as a hyperelastic Mooney-Rivlin material. Coefficients $C_1$ and $C_2$ for articular cartilage were assumed to be 4.1 MPa and 0.41 MPa, respectively, with $\nu$=0.4 [6]. A FE coordinate system was created from the polygonal surface of the reconstructed registration block. A corresponding coordinate system was established for the experimental measurements using the digitized coordinates of the registration block [2]. Nodes superior to the iliac guide wires and nodes along the pubis synthesis joint were constrained to simulate the boundary conditions in the experiment. Frictionless contact was enforced between the femoral implant and cartilage. Load was applied to the implant using the same magnitude and direction measured experimentally. All analyses were performed with the implicit time integration capabilities of LS-DYNA (Livermore Software Technology Corporation, Livermore, CA). FE predictions of cortical principal strains were averaged over elements that were located beneath each strain gauge. Best-fit lines and $r^2$ values were reported for each model at all loads for purposes of FE validation and sensitivity analysis.

*Sensitivity studies.* Sensitivity studies were performed to assess the effects of variations in assumed and measured material properties and cortical thickness on the predicted cortical surface strains (Table 1).

Table 1: Models studied for FE sensitivity analysis. Standard deviations (SD) in material properties and cortical thickness were taken from experimentally measured values (EXP) as well as deviations and ranges of values reported in the literature.

| Sensitivty Models Analyzed | Reference |
|---|---|
| Constant cortical bone thickness = ± 0, 1 SD (0.44 mm) | EXP |
| Constant trabecular bone modulus, $E$ = ± 0, 1 SD (80 MPa) | EXP |
| Constant cortical thickness & trabecular modulus , 1.0 mm, $E$ = 160 MPa | EXP |
| Trabecular bone Poisson's ratio, ν=0.2, ν=0.39 | [7] |
| Cortical bone Poisson's ratio, ν=0.29 | [8] |
| Cortical bone modulus, $E$ = ± 1 SD (1.62 GPa) | [9] |
| Cartilage thickness = 0.0, 4.0 mm (Min/Max) | EXP |
| Cartilage modulus, $E$ = 1.36, 7.79 MPa (Min/Max) | [10] |

RESULTS

*Cortical Thickness, Trabecular Elastic Modulus*. Using boxes, spheres, and parallel planes with known thickness, it was determined that the maximum RMS error of the thickness algorithm was ±2%. Cortical bone thickness ranged from 0.44 to 3.83 mm (mean 1.0 ± 0.4 mm) for the cadaveric pelvis (Fig. 6). Using empirical relationships between CT Houndsfield unit, apparent density and elastic modulus, it was determined that trabecular elastic modulus ranged from 20-400 MPa (mean 160±80 MPa).

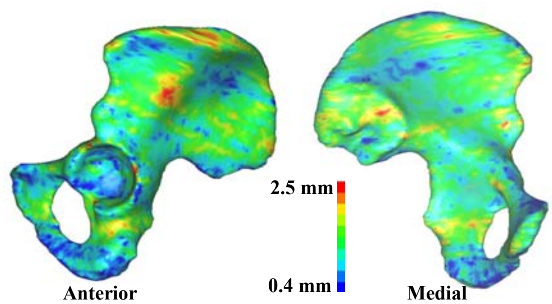

Figure 6: Contours of position dependent cortical bone thickness. Cortical thickness was highest along the iliac crest, the ascending pubis ramus, at the gluteal surface and around the acetabular rim. Cortical bone was minimal at the acetabular cup, the ischial tuberosity, the iliac fossa and the area surrounding the pubic tubercle.

*FE Model Predictions*. The subject-specific FE model predictions of principal strains showed strong correlation ($r^2$=0.776) with experimental measurements (Fig. 7, top panel). Similar correlation coefficients were obtained for the models examined in the sensitivity studies. Models representing standard deviations in average trabecular elastic modulus did not alter strains considerably (Fig 7, middle panel). In contrast, changes in the thickness of the cortical bone had a substantial effect on cortical strains (Fig 7, bottom panel). Nevertheless, the model with average cortical thickness predicted strains that were consistent with subject-specific model results (data not shown). Alterations to all other material properties did not alter cortical strains significantly (data not shown).

DISCUSSION

A subject-specific finite element model of the pelvis was developed and validated. Accurate FE model predictions were obtained when position-dependent cortical thickness and elastic modulus were used. The FE models were most sensitive to alterations in cortical bone thickness. Accurate predictions of cortical strains were obtained when an average cortical thickness and modulus was used; thus, the subject-specific model was not any more accurate than those that assume average properties. However, the average cortical shell thickness in areas where strain gauges were present was very close to the average thickness for the entire model but deviated much less (mean 1.07±0.2 mm). This suggests that accurate measurement of cortical thickness should be considered for future models since the model predictions were very sensitive to deviations in this input.

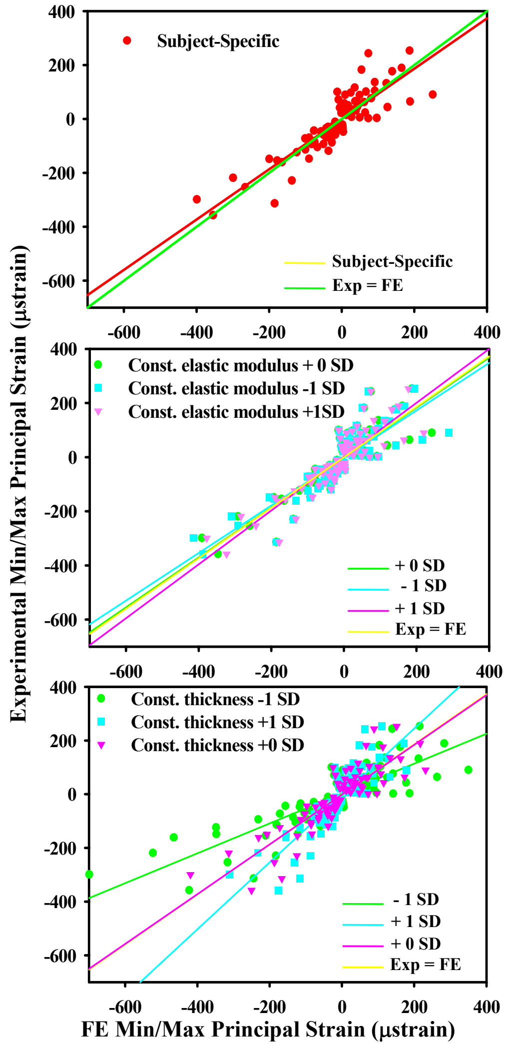

Figure 7: FE predicted vs. experimental cortical bone principal strains. Top panel- subject-specific, middle panel- constant trabecular modulus, bottom panel- constant cortical thickness.

Three-dimensional FE models of the pelvis have been studied on several occasions. Most of these models used simplified pelvic geometry, average material properties and did not validate FE predictions of stress and strain. To our knowledge, the work of Dalstra et al. was the first and only attempt to develop and validate a three-dimensional FE model of the pelvis using subject-specific geometry and material properties [3]. The FE model was validated using experimental measures of strain in the peri-acetabular region of a cadaveric pelvis. However, validation by direct comparison with subject-specific experimental measurements was not performed. Different cadaveric specimens were used for FE mesh generation and experimental tests.

The results of the sensitivity studies demonstrated the relative importance of both assumed and measured parameters for FE modeling of the pelvis. This study showed that the prediction of cortical bone strains was most sensitive to changes in cortical thickness. Alterations to both bone Poisson's ratios, cartilage elastic modulus, and cartilage thickness did not alter cortical strains. However, changes to these inputs did have an effect on overall model displacement and cartilage/trabecular bone mechanics. Therefore, model sensitivity should not be approached as "all or none"; the required accuracy of individual material inputs/parameters will depend on the nature of the study.

Although the results of the sensitivity studies suggest that errors in material property values will not likely to produce significant changes in the stress and strain gradients in the cortical bone, it is likely that strains would be more sensitive to changes in the boundary conditions and applied loading conditions. For this reason, it was not the intent of this study to replicate physiological loading conditions. By applying a well-defined experimental loading configuration, the applied loads were accurately replicated in the FE model with minimal error.

It is concluded that the presently developed approach for subject-specific FE modeling of the pelvis has the ability to predict cortical bone strains accurately. The techniques validated in this study can be applied to the analysis of the detailed pelvic geometry of individual patients based on volumetric CT scans. This will provide a means to examine the mechanics of the pelvis for cases when detailed subject-specific geometry should be considered.

## ACKNOWLEDGMENTS

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