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Development of a Computer Simulation Tool for Application in Adolescent Spinal Deformity Surgery

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Abstract. Scoliosis is a three-dimensional spinal deformity which requires surgical correction in progressive cases. In order to optimize correction and avoid complications following scoliosis surgery, patient-specific finite element models (FEM) are being developed and validated by our group. In this paper, the modeling methodology is described and two clinically relevant load cases are simulated for a single patient. Firstly, a pre-operative patient flexibility assessment, the fulcrum bending radiograph, is simulated to assess the model's ability to represent spine flexibility. Secondly, intra-operative forces during single rod anterior correction are simulated. Clinically, the patient had an initial Cobb angle of 44 degrees, which reduced to 26 degrees during fulcrum bending. Surgically, the coronal deformity corrected to 14 degrees. The simulated initial Cobb angle was 40 degrees, which reduced to 23 degrees following the fulcrum bending load case. The simulated surgical procedure corrected the coronal deformity to 14 degrees. The computed results for the patient-specific FEM are within the accepted clinical Cobb measuring error of 5 degrees, suggested that this modeling methodology is capable of capturing the biomechanical behaviour of a scoliotic human spine during anterior corrective surgery.

Keywords: anterior scoliosis surgery, spinal deformity, patient-specific finite element model, surgery simulation

1 Introduction

Adolescent idiopathic scoliosis (AIS) is the most common spinal deformity, and requires surgical correction in progressive cases. Corrective surgery aims to reduce the abnormal spinal curvature and prevent further progression of the deformity via removal of the intervertebral discs, insertion of bone graft material into the cleaned disc spaces and attachment of metal rods to the spine using screws. Post-operative complications (such as screw pullout) or suboptimal correction can occur due to inappropriate choice of surgical levels or the application of excessive corrective force during the procedure. Biomechanical computer models of the spine have the potential to help optimise surgery outcomes and reduce complications, and patient-specific finite element (FE) models have been utilized previously to investigate the biomechanics of AIS surgery [1, 2]. The current study aims to develop more anatomically detailed FE models of scoliosis patients, for subject-specific prediction of the loading and deformation of individual spinal structures (eg ligaments and implants) during surgery. Such mechanical data would provide an improved ability to predict surgical outcomes. An important part of the model development process is the validation of model predictions by comparison with clinical data, and an initial model validation for a single patient is the subject of this paper.

2 Methods

Purpose developed image processing and FE pre-processing tools were developed to allow rapid generation of subject-specific FE models of patients from low-dose pre-operative computed tomography (CT) datasets. As an initial step in validation of these computational tools, they were used to create a subject-specific model of a single scoliosis patient, and the model predictions for two clinically relevant physiological load cases were compared with clinical data for this patient. All analyses were performed on a HP xw660 workstation (Intel Xeon 5420, 4GB RAM) using Abaqus/Standard 6.7.1 (Simulia Inc, RI). Analyses were quasi-static with non-linear (finite strain) geometry capability enabled.

2.1 Patient-Specific FE Model Geometry for the Intact Spine

Our method for deriving patient-specific FE models from CT scan data has been previously described [3]. The three-dimensional, pre-operative CT dataset for an AIS patient (14yo female, 65kg, 165cm, pre-op major Cobb angle 44^{o}, Lenke Class 1A) was imported into a custom developed image processing software (Matlab R2007b, The Mathworks, Natick, MA) where the osseous anatomy was thresholded and key bony landmarks were manually selected by the user. These landmarks were imported to a custom FE pre-processing tool (Python 2.5) which generated a parametric FE model of the osseoligamentous thoracolumbar spine, including vertebrae, ribs, sternum, discs, joints, and ligaments (Figure 1a). Seven spinal ligaments were simulated at each vertebral level (Table 1) and these were simulated as either linear connections, or in the case of the anterior and posterior longitudinal ligaments, as a group of spring elements in series and parallel. Note that while the model anatomy is patient specific (derived from CT scan data), all tissue material properties for the spine model are derived from existing literature. Details of the element types and material properties used are provided in Table 1.

Table 1. Element representations and material properties used in the subject-specific FE model of the thoracolumbar spine

Anatomy	Element type	Material properties	Ref
Bony anatomy			
Cortical bone	3D, 4-node shell	Linear Elastic E = 11.3GPa; ν = 0.2	[4]
Cancellous bone	3D, 8-node brick	Linear Elastic E = 140MPa; ν = 0.2	[4]
Posterior elements	3D, 2-node beams	Quasi-rigid	
Intervertebral discs			
Intervertebral disc anulus ground matrix	3D, 8-node brick	Hyperelastic (Mooney-Rivlin) C_{10} = 0.7; C_{01} = 0.2	[5]
Intervertebral disc collagen fibres	3D, tension-only link (embedded rebar)	Linear elastic: E = 500MPa; ν = 0.3	[6]
Intervertebral disc nucleus pulposus	3D, 4-node hydrostatic fluid	Incompressible fluid: E ≈ 0, ; ν = 0.5	[7]
Cartilaginous joints			
Zygapophyseal joint surfaces	3D, 4-node shell	Linear Elastic E = 11.3GPa; ν = 0.2	
Costovertebral joints	3D, 2-node beams	Linear Elastic E_{compr} = 245N.mm^{-1} $k_{torsion}$ = 4167Nmm.rad^{-1} $k_{bending}$ = 6706Nmm.rad^{-1})	[8]
Ligaments			
Ligamentum flava, supra-/inter-spinous, capsular, intertransverse	3D, 2-node, tension-only connectors	Piecewise, non-linear elastic	[9, 10]
Anterior/posterior longitudinal ligament	3D, 2-node spring	Piecewise, non-linear elastic	[10]
Implant construct			
Rod	3D, 8-node brick	Linear elastic, perfectly plastic: E = 108GPa; ν = 0.3; σ_y = 390MPa	
Screws	3D, 8-node brick & 3D, 2-node beams	Same as for the rod	

2.2 Simulating Surgically Altered Spinal Anatomy

The AIS patient represented in this study underwent a single rod, anterior corrective procedure, with vertebral screws at levels T5 to T12 and discectomies at levels T5-6 to T11-12. The custom pre-processing software was capable of automatically re-generating the surgically altered geometry and FE mesh using user-defined details for the screw location/orientation, discectomy levels and rod size (Figure 1b). The simulated discectomies were represented by removal of half the anulus mesh, and removal of the entire incompressible, fluid filled cavity representing the nucleus pulposus. Contact between the exposed surfaces of the adjacent vertebrae was simulated using both an exponential, softened contact algorithm (normal contact) and a Coulomb friction model, μ=0.3 (tangential contact). Vertebral screws were represented in the vertebral bodies between T5 and T12. The idealized screw shaft representation simulated a perfectly bonded relationship between the screw surface and the underlying cancellous bone elements, without consideration of the screw threads embedded within the bone.

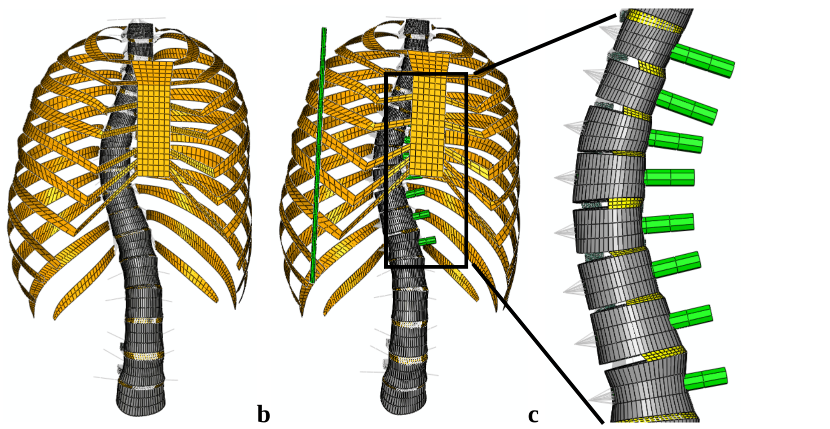

Fig. 1. Finite element mesh a. Intact scoliotic spine; b. Surgically altered spinal geometry with the rod and screws shown in green; c. Surgically altered region of the simulated spine, showing the remaining disc annulus elements (yellow) and screws with beam elements (grey wire) simulating the screw heads (screw ends have been lengthened to visualize screw positioning)

2.3 Simulated Load Cases in the Intact and Surgically Altered Spine

Two loading cases were simulated; (i) pre-operative fulcrum bending test on the intact spine (*intact* model) and (ii) intra-operative implant positioning on the surgically altered spine (*surgery* model).

1. Pre-operative fulcrum bending radiograph. Scoliosis patient spinal flexibility is often assessed clinically using a fulcrum bending radiograph [11] whereby the patient lays laterally over a cylindrical bolster, such that the convex side of their curve is adjacent to the bolster surface. This is a passive correction, which is not driven by muscle activation. To simulate this activity in the *intact* model, patient specific CT-derived segmental torso weights for each vertebral level were determined using custom-developed software (Matlab 2007b) and applied at the centroid of the transverse CT slice corresponding to that vertebral level (tissue density = 1.04×10^{-3} $g.mm^{-3}$). Additionally, load vectors simulating the weight of the full left arm and the upper portion of the right arm were applied at the T1 centroid and a load vector representing the weight of the head/neck was simulated as a point load superior to the T1 vertebra [12] (Figure 2). The bolster was modelled as a rigid body and rigidly constrained. A frictionless contact relationship existed between the ribs and the bolster surface and the spine was free to rotate about a point simulating the contact between the pelvis and the table. Rigid body rotation of the model is prevented by the combination of the translational constraint at the simulated point of contact between the pelvis and the table, and the positive contact pressure between the spine model and the stationary bolster under simulated gravitational loading.

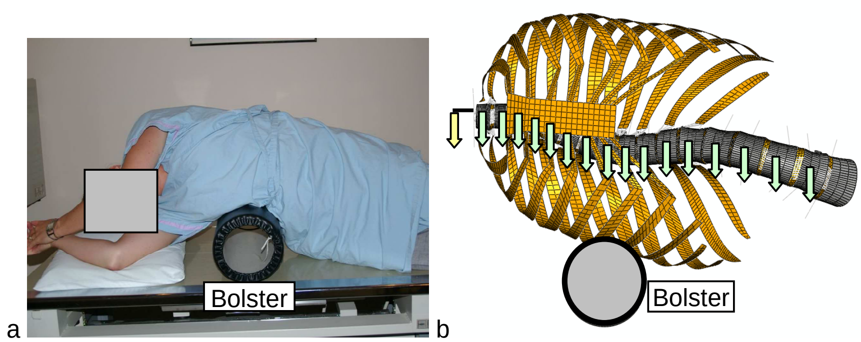

Fig. 2. a. Patient in position for the fulcrum bending radiograph. b. *intact* model after simulating the fulcrum bending activity. Green arrows = segmental vertebral torso weights; Yellow arrow = Head weight

The simulated Cobb angle in the unloaded *intact* model was compared with the pre-operative Cobb angle measured clinically from standing radiographs. The Fulcrum Flexibility (FF[1]) is used clinically to characterise a patients' flexibility during the fulcrum bending test. This parameter was calculated for the simulated deformed shape of the spinal column in the *intact* model and compared with the clinically measured value. An error of 5^o between clinical and simulated Cobb angle measurements was considered acceptable since this is the clinically accepted value for accuracy in Cobb angle measurements.

[1] Change in Cobb angle between the standing and fulcrum bending radiographs, expressed as a percentage of the standing Cobb angle

2. Intra-operative implant insertion. Following removal of the discs and insertion of the screws, the anterior surgical procedure involves successive compression of the intervertebral joints within the structural curve. The compressive force is applied between screw heads at adjacent vertebrae, thus resulting in a successive, level-wise decrease in the overall deformity. Data for the corrective forces applied intra-operatively were obtained *in situ* by Cunningham *et al* [13]. These data were utilized in the *surgery* model to simulate the incremental, level-wise compression of adjacent vertebral joints in the thoracic spine. The L5 vertebra was constrained from motion during all load steps. The simulated corrected Cobb angle was compared with the clinically measured corrected Cobb angle obtained immediately post-operative, to ascertain the accuracy of the *surgery* model in predicting the change in coronal deformity following surgery (acceptable error = 5^o).

3 Results

3.1 Fulcrum Bending Load Case

Clinically, this patient demonstrated an initial Cobb angle of 44^o, which reduced to 26^o during Fulcrum Bending, thus giving a pre-operative clinical Fulcrum Flexibility of 40.9%. The simulated initial Cobb angle was 31^o, however, it has been shown that the standing Cobb angle measurement for AIS patients was on average 9^o higher than the Cobb angle measured while the patient lays supine [14]. As such, the simulated, corrected-Cobb angle was 40^o, which was within the accepted angular measurement error of 5^o (Figure 3). The simulated Cobb angle reduced to 23^o when the displacement of the spine during the Fulcrum bending radiograph was simulated. This was within 5^o of the clinically measured value. Thus the simulated FF was 47.8%, which was calculated using the simulated deformed Cobb angle and the clinically measured standing Cobb angle (Figure 3).

3.2 Intra-Operative Load Case

Following surgery, the clinically measured Cobb angle was 14^o – this was measured one week post-operative. Results from the surgery simulation demonstrated a simulated corrected Cobb angle of 14^o (Figure 3). As such, the computed results for the patient-specific FEM were the same as the clinical data.

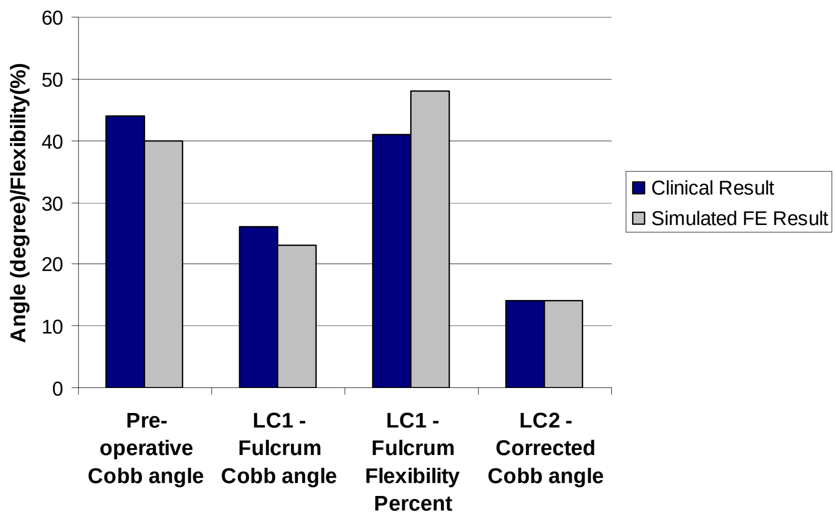

Fig. 3. Comparison of clinical and simulated results for a single AIS patient analysed under the fulcrum bending load case (LC1) and intra-operative correction load case (LC2)

4 Discussion

This paper presents the preliminary validation of a computational tool for developing subject-specific FE models of scoliosis patients, both prior to and immediately following corrective surgery. A patient-specific FE model for a single AIS patient was analyzed for physiological load cases representing both a pre-operative loading condition on the intact spine and an intra-operative loading condition representing intra-operatively applied surgical corrective forces. Comparison between the clinical and simulated data for both the *intact* and the *surgery* models demonstrated good agreement (within 5^o error), suggesting that the patient-specific modeling capabilities hereto developed are capable of capturing the physiological behaviour of a scoliotic spine.

We note that our use of a 5^o criteria for comparison of model predictions with radiographic measurements does not imply that the model results cannot be resolved more finely than 5^o, nor that the model is not sensitive to changes of less than 5^o. Rather, it is well known that radiographic Cobb angle measurements in scoliosis vary by around 5^o due to inter and intra-observer error [15], so this comparison range is necessitated by the uncertainty in the radiographic measurements used for model comparison.

While these patient-specific modeling capabilities currently do not allow the inclusion of patient-specific muscle forces, arguably the physiological loading conditions for which the models have been validated do not include muscle activation. Additionally, this preliminary model validation was carried out using material parameters derived from the literature, which are exclusively for adult spinal tissues. Ideally, material parameters derived for paediatric tissue would be incorporated in the model, however, such data is not available in the literature.

Biomechanically, the fulcrum bending test provides a potentially attractive clinical assessment tool which could be used to help prescribe patient-specific soft tissue properties in the spine model. By adjusting soft tissue properties until the model fulcrum flexibility matches the clinical measured value, the model could be 'calibrated' to match the soft tissue properties of a particular patient. However, there are several difficulties with this approach. Firstly, spinal flexibility is governed not by one soft tissue structure, but by a combination of discs, ligaments, and facet joints interacting in a manner unique to the loading being applied, therefore adjusting the mechanical properties of seven ligaments as well as the intervertebral disc and facet joints to match a single test value (the fulcrum flexibility) will not provide a unique solution to the problem of inversely determining soft tissue properties. Secondly, the manner in which soft tissue properties affect fulcrum flexibility is currently not clear. A previous study by our group using the same model [3] found that reductions of up to 40% in disc collagen fibre stiffness and ligament stiffness produced no measurable increase in fulcrum flexibility. However, complete discectomy did provide a large increase in simulated fulcrum flexibility, suggesting that the discs play an important role in governing fulcrum flexibility.

Future validation studies will develop upon the preliminary validation presented here, to use a larger subset of patient data, thus providing a more detailed and thorough validation of the patient-specific spine FE models. Using patient-specific FEM it will be possible to gain an improved understanding of the biomechanical impact of surgical interventions on the structures within the spine. Many of the complications associated with scoliosis corrective surgery are mechanical in nature and use of a computational tool such as this will provide surgeons with an improved ability to predict the likely outcome following scoliosis surgery.

References

1. Aubin, C.E., et al., *Biomechanical modeling of posterior instrumentation of the scoliotic spine.* Comp Meth Biomech Biomed Eng, 2003. **6**(1): p. 27-32.
2. Dumas, R., et al., *Personalized body segment parameters from biplanar low-dose radiography.* IEEE Trans Biomed Eng, 2005. **52**(10): p. 1756-63.
3. Little, J.P. and C.J. Adam, *The effect of soft tissue properties on spinal flexibility in scoliosis: biomechanical simulation of fulcrum bending.* Spine (Phila Pa 1976), 2009. **34**(2): p. E76-82.
4. Lu, Y.M., W.C. Hutton, and V.M. Gharpuray, *Do bending, twisting, and diurnal fluid changes in the disc affect the propensity to prolapse? A viscoelastic finite element model.* Spine, 1996. **21**(22): p. 2570-9.
5. Little, J.P., *Finite element modelling of anular lesions in the lumbar intervertebral disc*, in *School of Mechanical, Manufacturing and Medical Engineering*. 2004, Queensland University of Technology: Brisbane, Australia. p. 361.
6. Kumaresan, S., et al., *Finite element modeling of the cervical spine: role of intervertebral disc under axial and eccentric loads.* Med Eng Phys, 1999. **21**(10): p. 689-700.
7. Nachemson, A., *Lumbar Intradiscal Pressure: experimental studies on post-mortem material.* Acta Orthopaedica Scandinavica, 1960. **43**.
8. Lemosse, D., et al., *Characterization of the mechanical behaviour parameters of the costo-vertebral joint.* Eur Spine J, 1998. **7**(1): p. 16-23.
9. Chazal, J., et al., *Biomechanical properties of spinal ligaments and a histological study of the supraspinal ligament in traction.* J Biomech, 1985. **18**(3): p. 167-76.
10. Nolte, L.P., M. Panjabi, and T. Oxland, *Biomechanical properties of lumbar spinal ligaments*, in *Clin implant mat*, G. Heimke, U. Soltesz, and A.J.C. Lee, Editors. 1990, Elsevier Science Publishing: Amsterdam. p. 663-668.
11. Cheung, K.M. and K.D. Luk, *Prediction of correction of scoliosis with use of the fulcrum bending radiograph.* J Bone Joint Surg, 1997. **79**(8): p. 1144-50.
12. Erdmann, W.S., *Geometric and inertial data of the trunk in adult males*. J Biomech, 1997. **30**(7): p. 679-88.
13. Cunningham, H., J.P. Little, and C.J. Adam, *The measurement of applied forces during anterior single rod correction of adolescent idiopathic scoliosis*, in *ACSR Annual Meeting*. 2009: Adelaide, Australia.
14. Torell, G., et al, *Standing and supine Cobb measures in girls with idiopathic scoliosis*. Spine, 1985. 10:425-7
15. Vrtovec, T., F. Pernus, B. Likar, *A review of methods for quantitative evaluation of spinal curvature*. Eur Spine J, 2009. 18:593-607.