**Supplemental Digital Content 1**

**Finite Element Model Development, Material Properties, Calculations, and Limitations**

Material in this supplement provides information regarding the model in addition to that provided Materials and Methods of the primary manuscript. We followed the applicable EQUATOR reporting guidelines for simulation.1

**Finite Element Model Development and Validation**

We used a previously developed finite element (FE) model of the complete human cervical spine, from the occiput (Oc) to the seventh cervical vertebrae (C7), and cervical spinal cord using ABAQUS v2020 software (Dassault Systèmes Simulia Corporation, Providence, RI). This model was first described in Gadomski, et al.2

The model incorporates aspects of two previously validated FE models of the occipital-cervical complex (Oc-C2)3 and sub-axial cervical spine (C3-C7).4,5 Three-dimensional bony geometry was obtained from a clinical resolution, quantitative computed tomography scan of a single, fresh-frozen, female cadaveric spine (age 64 y, height 170 cm, weight 74 kg).6 The biomechanical representation and material properties of bone (cortical, trabecular, endplate), intervertebral discs (annulus fibrosus, nucleus pulposus), articular cartilage, and 17 ligaments and membranes have been described in detail; 6 see table S1-1.7-16 A full convergence study of each tissue simulated in the model using strain energy criteria established the requisite degree of mesh refinement.6

**Table S1-1**. Material Mechanical Properties Used in the Finite Element Model

|  |  |  |  |
| --- | --- | --- | --- |
| **Structure** | **Reference** | **Elastic Modulus (MPa)** | **Poisson Ratio (ν)** |
| Cortical bone | Ueno and Liu, 19877 | 11,000.00 | 0.30 |
| Trabecular bone\* | Crawford et al., 20038 |  |  |
| Bony endplates | Whyne et al., 20019 | 1,000.00 | 0.30 |
| Articular cartilage† | Noailly et al., 200510 |  |  |
| Annulus fibrosus‡ | Fagan et al., 2002,11 Natarajan et al., 200412 | 5.00 | 0.40 |
| Nucleus pulposus‡ | 1.50 | 0.49 |
| Spinal ligaments§ | Yoganandan et al., 2001,13 Rohlmann et al., 200614 |  |  |
| Posterior elements | Dooris et al., 200115 |  |  |
| Spinal cord∥ | Sparry and Keaveny, 200916 |  |  |

\* Houndsfield attenuation data from the original CT scan were converted to spatially varying orthotropic elastic moduli values for the vertebral body trabecular bone.

† Marlow initial bulk modulus = 1.80 MPa; initial ν =0.40.

‡ The annulus fibrosus of the intervertebral discs was defined using a linear elastic material definition, and the nuclei were defined as a nearly incompressible (ν= 0.49) isotropic material.

§ The spinal ligaments were simulated as nonlinear spring elements, where the reaction force (F) was related to engineering strain (ε), material constants (a, b), and the stress-free state constant (c):14

∥ Experimentally-derived and published material coefficients (μp=0.45 and αp=4.70) were used to describe the spinal cord’s finite deformation material behavior with an Ogden first-order hyperelastic strain energy function expressed as a function of finite strain principal stretches ():16

Published human cross-sectional geometry data were used to model the cervical spinal cord.17 Experimentally derived material coefficients were used to describe the spinal cord’s finite deformation material behavior.16

FE model predictions of segmental motion (axial rotation, lateral bending, flexion/extension) in the intact spine were in close agreement (mean plus 1 SD) with those measured experimentally in whole Oc-C7 cadaveric spines undergoing pure moment loading. 6  Likewise, experimentally measured facet contact forces at C1-C2 were comparable to model predictions. Additional validation studies were conducted with cadaveric specimens with ligamentous injuries at C5-C6. With pure moment loading, model and experimental values for three motions at the injured C5-C6 segment were in close agreement.6  This FE model has been shown to accurately predict cervical spine intervertebral motion (rotation, translation) in response to applied intubation forces. Specifically, experimentally measured laryngoscope forces were utilized to model cervical spine motions in an intact (stable) cervical spine in living patients, and with a C2 fracture (Type II odontoid fracture) and severe C3-C4 distractive-flexion injury in cadavers.2

The model consists of 196,984 elements, 237,635 nodes, and 671,997 degrees of freedom. For this study, all ABAQUS jobs were computed on a Linux operating system (CentOS Linux 7) with 60 processor cores (Xeon E5 2683 v4, Intel Corporation, Santa Clara, CA) and 128 GB RAM. In all simulations, force magnitude was applied incrementally using minimum and maximum increments of 1x10-7% and 1% of the total force, respectively, depending on the complexity of the simulation until converging on the assigned force magnitude (100%). The minimum simulation runtime was 1.1 hours and the maximum simulation runtime was 9.9 hours. Resultant motion and strain values represent quasi-static values corresponding to the maximum values occurring during intubation

**Strain Calculations**

*Ex vivo* studies indicate accounting for strain (ε) in multiple simultaneous planes (dorsoventral [εYY], mediolateral [εXX]), and shear strain [εXY]) correlates better with axonal injury than strain in any single plane.18 Therefore, we utilized two measures of strain that incorporate the overall strain field without consideration for direction: 1) *maximum* principal strain (εMAX, analogous to *stretch*) and 2) *minimum* principal strain (εMIN, analogous to *compression*). These principal strain values were calculated using the following equation where the three roots of the strain cubic equation (ε) are the principal strains:

and I1, I2, and I3 are the first three invariants of the strain tensor:19

**Model Limitations**

First, when compared with patients, the current FE model appears to underestimate intubation-mediated extension at C3-C4 and C4-C5.2 This may be due, at least in part, to the imposed kinematic constraint of the C7 vertebral body. Although the difference between observed and predicted motion is quantitatively small, we cannot estimate how much this difference affects modeled cord strains. In a future version of this FE model, inclusion of an additional caudal vertebral segment (e.g., T1) will permit C7-T1 motion, and this may increase modeled sub-axial motion.

Second, the current FE model does not include an explicit representation of spinal cord gray and white matter. These tissues may20,21 or may not22 have different primary biomechanical properties and the rostral-caudal alignment of axonal fibers in the spinal cord white matter provides a direction-specific mechanical response.21 Gray matter may have lesser strain tolerances than white matter,23-27 although the difference is relatively small (10-20%). Thus, modeled spinal cord strain fields28,29 and regional (intra-cord) and individual susceptibility to strain injury are likely to be more complex than are represented in the current version of our FE model.

Third, although most of the materials implemented in the FE model were defined with nonlinear material properties, several material assignment simplifications were required for computational purposes, most notably the linear elastic material definition assigned to the annulus fibrosis. While these simplifications directly affect model predictions of the localized mechanical environment of the disc (not germane to the current study), they do not alter the spinal motions nor do they effect predictions of spinal cord strain.

Fourth the current FE model does not include a representation of cerebrospinal fluid, which has been shown to affect FE model predictions of spinal cord compression under dynamic loading conditions (with the CSF providing a dampening or “cushioning” effect). However, because quasi-static loading conditions were used in the current study, the absence of CSF is not likely to substantially affect the reported results. The inclusion of CSF in future modeling efforts will allow for predictions of transient changes in spinal cord stresses and strains, wherein both the magnitude and duration of supraphysiological cord compression has been shown to play a key role in the resulting acute and permanent neurological impairment.30

Fifth, because model anatomy was derived from a single cadaver6 and mean material property data inputs were utilized to define the model, the lack of geometric and material property input variation produces deterministic (*i.e*., single-valued) motion and strain values. Accordingly, the current FE model does not simulate the inherent variation across the human population but, instead, represents an anthropometrical mean, i.e., an average patient. In the future, to account for the variation in both geometry and material properties across the general population, probabilistic methods will be dovetailed with the current FE model. The underlying principle of probabilistic analyses is that the input parameters (anatomy, material properties) are not defined by a single value but are strategically sampled from a distribution that represents the population’s variation and the model is solved many times to develop a distribution of the output variables of interest.31

Sixth, this model does not distinguish between male and females. Recent studies of head, neck, and cervical spine morphology32-35 and tissue material properties36,37 indicate male and female cervical spines are significantly different.38  Overall, the male cervical spine is significantly less mobile than that of the female. This may be the basis for why intubation forces are greater in males than in females, independent of body weight.39 Sex-specific differences in spine morphology, tissue properties, and intubation biomechanics may result in sex-specific differences in intubation-mediated cervical spine motion and cord strains.

Seventh, the current model is based on an anatomically normal cervical spine and spinal cord. Future variations of the FE model may be able to replicate cervical spine degenerative disease (spondylosis/stenosis).41-43  This may allow modeling of cord strains during intubation and/or other dynamic motion conditions in the presence of chronic cord compression.

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