Sensitivity analysis of a morphological finite element L3-L4 FSU to assess biomechanical responses

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Abstract The goal of this study is to investigate the influence of morphological variations in the responses of a single lumbar vertebral segment under quasi-static moment loading. A finite element (FE) model of L3-L4 functional spinal unit (FSU) with parametric anatomical variations in disc wedge angle (DWA), transverse diameter (TD), vertebral body height (VH), and anteroposterior diameter (APD) is used to generate a population space with constrained lattice hypercube sampling algorithm. Three biomechanical responses were analysed for each sample: rotation angle (RA), change in disc height (c_DH), and disc pressure (DP). At 10 Nm moment, RA ranged between 3.68° and 9.64° and was highly influenced by APD, while least sensitive to DWA. A Bayesian regression model was used to compute the effect of each anatomical parameter and to predict the RA. The Bayesian approach, developed using a FE population study, promises a reliable and cost-effective way of predicting biomechanical responses in the FSU considered in this study and could potentially be extended to complete lumbosacral spine.

Keywords L3-L4 FE FSU, sensitivity analysis, parametric modeling, population study, Bayesian regression model.

I. INTRODUCTION

Lumbar spine plays a crucial role in human biomechanics, serving as a primary structure for load transmission, movement, and stability in the body. It resists shear forces, bears compressive loads, and facilitates motions such as flexion, extension, and lateral bending [1-3]. To understand the biomechanics of the lumbar column, functional spinal units (FSUs) are often studied as representative models of the overall behaviour of the spine. Among these, the L3-L4 FSU is frequently analysed due to its representative biomechanical characteristics and capacity to endure tensile and shear deformations during various activities [3]. Degenerative changes or instrumentation in FSUs such as L3-L4 can alter load-sharing mechanisms and increase stress on adjacent levels, which may lead to conditions like adjacent segment degeneration (ASD) [4]. Biomechanical responses such as Rotation Angle (RA) and stress distribution within this segment provide important insights into spinal health, enabling experimental and computational studies to explore the effects of different loading and pathological conditions.

Despite advancements in finite element (FE) modeling for the lumbar spine, a major challenge lies in accounting for the variability of anthropometric parameters across populations. Parameters such as disc wedge angle (DWA), transverse diameter (TD), vertebral height (VH), and anteroposterior diameter (APD) influence biomechanical responses, including (RA) and disc pressure (DP) [1-2]. Population-based parametric modeling using morphological frameworks provides a robust method to capture these variations and to analyse the mechanical behaviour of FSUs like L3-L4 under diverse loading conditions.

This study aims to investigate the biomechanical behaviour of the L3-L4 FSU by employing morphological FE modeling and Bayesian regression analysis. By simulating variability in anatomical parameters, this work seeks to offer insights into the biomechanical implications of the L3-L4 segment, informing personalised diagnostics and treatment strategies and understanding crash injury risks.

II. METHODS

The L3-L4 FSU used in this work was extracted from a 50th percentile validated female morphological lumbar spine model [5]. The model comprised 5,912 hexahedral solid elements, 3,450 quad shell elements, and 20 beam elements. The mesh quality was evaluated based on aspect ratio, skewness, and Jacobian. The thresholds were set as follows: aspect ratio < 5 and < 8 for shell and solid elements, respectively; skewness < 50°; and Jacobian > 0.5. The cortical and trabecular bone of the L3-L4 FSU were modeled using an isotropic linear elastic material model [6]. The intervertebral disc, consisting of the annular ground substance, annulus fibrosus, and nucleus pulposus, was modeled using Hill foam [6], fabric [7], and viscoelastic [8] constitutive material models, respectively. Capsular ligaments were modeled as beam elements, while all other ligaments in the FSU were

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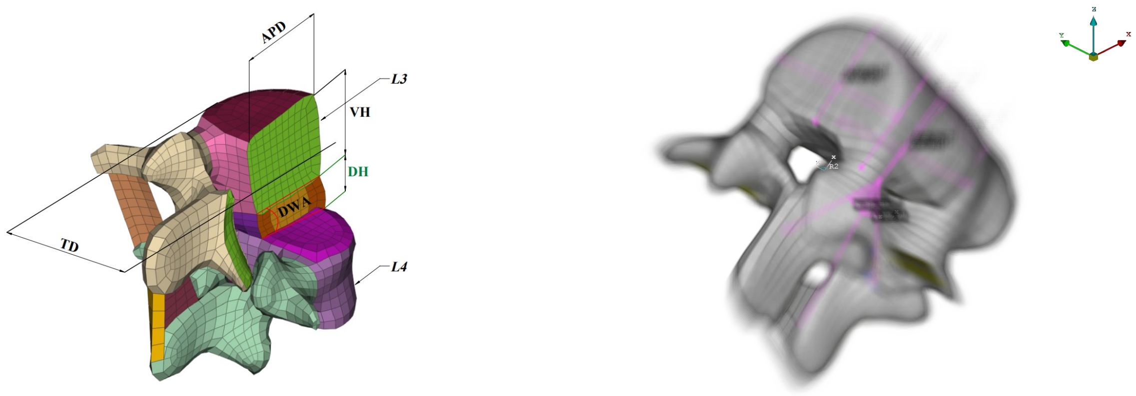
modeled using a fabric material model with non-linear stress-strain curves [9-10]. Facet joint contacts were established using the surface-to-surface contact elements. The lower endplate of the inferior vertebra was constrained, and a quasi-static moment load of 10 Nm was applied to the upper rigid endplate of the superior vertebra. The rotation angle in the sagittal plane, i.e. RA for flexion and extension, was considered the primary metric for validating the model.

Model Parameterisation and Simulation Setup

The model was parameterised to generate new FE models with five parametric variations (Fig. 1(a)). Anatomical parameters such as Disc Wedge Angle (DWA), Transverse Diameter (TD), Vertebral Body Height (VH), and Antero-Posterior Diameter (APD) were selected due to their influence on response variables such as flexion-extension RA, Disc Pressure (DP) and change in Disc Height (c_DH) [11]. Based on these anatomical parameters, directional constraints and specific variable control entities were defined for the model using the direct morphing method using ANSA BETA CAE™. To maintain the boundary geometry and better mesh quality of the intervertebral disc, TD and APD were chosen as global variations involving the entire FSU, while VH was chosen as a local variation involving a single vertebra.

There were four linear parameters (TD, APD, VH_L3, VH_L4) and one angular parameter (DWA). A population of 250 samples for each loading mode was generated using Constrained Latin Hypercube Sampling (CLHS) algorithm (Fig. 1(b)). The range of variation for DWA and scaling factors for VH, TD and APD are given in Table I. ANSA v23.1.1® was used for model generation, and the LS-DYNA R14.1 SMP® solver was employed to perform the simulations.



(a) (b)

Fig. 1. Morphological FE model of L3-L4 FSU: (a) L3-L4 FSU labelled with anatomical parameters, (b) transparent images of FSU population combined.

TABLE I

ANATOMICAL PARAMETERS WITH RANGE, BASELINE DIMENSIONS AND REFERENCES (UNITS: MM AND DEGREES)

Parameter	Range	Baseline	Lower Bound	Upper Bound	Reference
DWA	±3°	5.43	2.43	8.43	9.0 ± 3.2 [12]
TD_L3	±10 %	43.55	39.20	47.91	39.7 – 54.9 [13]
TD_L4	±10 %	45.30	40.77	49.83	–
VH_L3	±10 %	29.07	26.16	31.98	27.9 ± 2.1 [14]
VH_L4	±10 %	25.47	22.92	28.02	27.4 ± 2.2 [14]
APD_L3	±10 %	29.76	26.78	32.74	28.3 – 35.9 [13]
APD_L4	±10 %	28.53	25.68	31.39	–

Model Outputs and Bayesian Regression Analysis

Three outputs — RA, DP, and c_DH – were extracted from each model by coupling LS-Prepost v4.7® and Python3® to avoid manual intervention. RA was computed as the change in angle of the superior endplate of the L3 vertebra with respect to its initial position. DH was measured from the anterior side in the sagittal plane. Bayesian multiple linear regression analysis was developed using Bambi and was used to determine the influence of anatomical parameters and loading mode on the RA.

The prior distribution and likelihood for the model are defined as follows:

$$\alpha \sim \mathcal{N}(6.86,\ 97.2),\ \beta_0 \sim \mathcal{N}(0.0,\ 48.1),\ \beta_1 \sim \mathcal{N}(0.0,\ 48.7),\ \beta_2 \sim \mathcal{N}(0.0,\ 1.62),$$

$$\beta_3 \sim \mathcal{N}(0.0,\ 48.8),\ \beta_4 \sim \mathcal{N}(0.0,\ 48.7),\ \beta_5 \sim \mathcal{N}(0.0,\ 5.62),\ \sigma \sim \mathcal{HT}(4,\ 1.12)$$

$$\mu = \alpha + \beta_0(\text{APD}) + \beta_1(\text{TD}) + \beta_2(\text{DWA}) + \beta_3(\text{VH-L3}) + \beta_4(\text{VH-L4}) + \beta_5(\text{Load})$$

$$\theta = \mu + \sigma$$

All the priors were normally distributed. The parameter α is the common intercept of the model, while β_{0-5} corresponds to slopes of the predictors APD, TD, DWA, VH_L3, VH_L4 and loading mode, respectively. σ is the common error term in the model. θ represents the posterior predictive, capturing the plausible range of RA for the given predictors. The Bayesian model was fitted using Markov Chain Monte Carlo (MCMC) No-U-Turn Sampler (NUTS) algorithm. Contrast distributions for the loading modes were computed using the posterior predictive distributions. The individual effects of each anatomical parameter (predictor) on RA are further analysed using the partial correlation coefficients, which were computed using design matrix statistics [15].

III. RESULTS

A population space of 250 samples was generated using CLHS algorithm with multifactor variations in the anthropometric bounds for each loading case. Uniform exploration of the parameter space is ensured by CLHS. Out of 500 simulations (flexion and extension of 250 samples), 496 simulations were completed successfully with normal termination These successfully completed simulations were used for further Bayesian model development and testing. The RA responses, along with c_DH and DP for each L3-L4 FSU versus the moment, are shown in Fig. 2. The experimental data are also superimposed (RA) for validation and better comparison [16-17].

The anatomical variations in 250 samples resulted in maximum RA of 9.64° in extension and 8.29° in flexion. Maximum DP ranges from 0.01 MPa to 0.63 MPa among the generated models. To maintain relative scale for plotting, absolute values of c_DH were taken. Maximum c_DH of 3.42 mm was observed in flexion (Exp_249).

Posterior predictive distributions of RA showed a reasonable difference between flexion and extension. In Fig. 3(a) the contrast plot indicates extension is always greater than flexion by an average of 1.32°. The distribution of partial correlation coefficients of biomechanical response with respect to the global and local anatomical parameters are provided in Fig. 3(b). The most influential parameter was APD (r=-0.64). While any change in DWA and TD had very negligible effect (|r|<0.15) on RA. Predictions were made for the new group of 11 FSUs by sampling from the posterior draws and comparing with the FE results in Fig. 3(c). The Bayesian model performance was further evaluated using mean squared error, which yielded a value of 0.311°, suggesting good fit of the model with the FE results.

IV. DISCUSSION

Based on the biomechanical population study of L3-L4 FSU, among the five variables APD showed higher correlation and was highly sensitive with the RA. Lesser change was seen in DH; the superior vertebra tends to rotate more, resulting in less pressure on the intervertebral disc. The DP variations among the population in this study were found to be similar when compared to other studies involving subject-specific models and in vivo studies [2][18]. The Bayesian approach for predicting the biomechanical response with very limited anthropometric variables can be an effective method for comparing interpersonal variations. Bayesian regression predicts the plausible range along with the uncertainties involved during prediction.

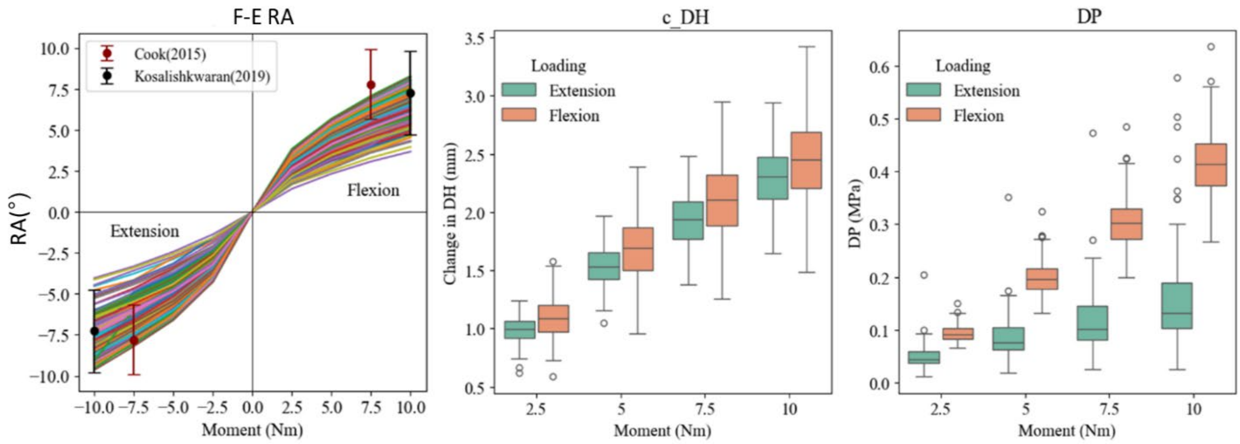


Fig. 2. Plots of biomechanical response (RA, c_DH and DP) from the population study at different moments.

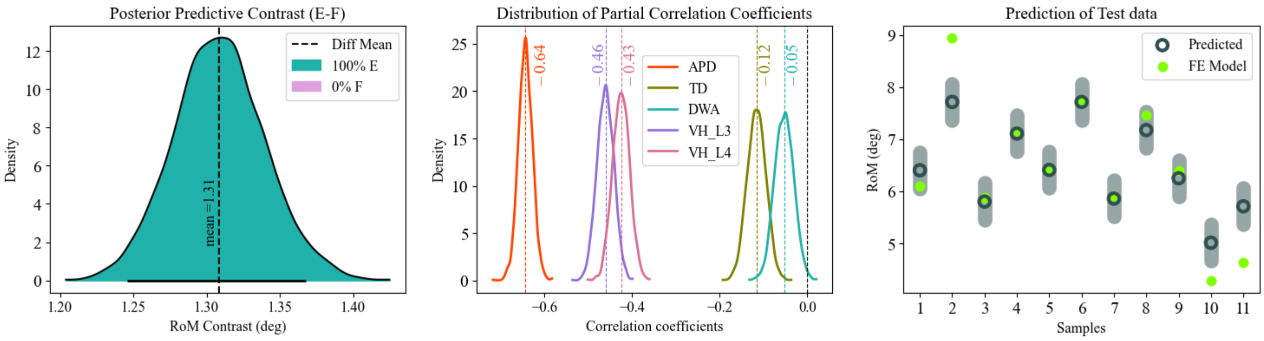


Fig. 3. Bayesian model outputs: (a) Left: distribution of contrast between extension and flexion, (b) Center: distribution of partial correlation coefficients of anatomical parameter with RA, and (c) Right: comparison of predicted plausible range of RA for test data with FE model results

V. CONCLUSION

The L3-L4 FSU was analysed for its sensitivity to RA under biomechanical loading conditions. Morphological variations in the anatomical parameters of the FSU influenced the RA and disc pressure responses. The RA was most influenced by the variations in the APD. The Bayesian model formulated from the FE population study serves as a highly reliable and cost-effective method for predicting biomechanical responses. The findings from this study enable us to associate the relation between RA and disc responses, which will eventually contribute to a better understanding of the biomechanics of the disc. Furthermore, these results can be utilised for analysing complications in surgeries like disc replacement, patient-specific clinical aftercare, and personalised crash injury outcomes.

VI. ACKNOWLEDGEMENTS

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