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# DEVELOPMENT OF A FINITE ELEMENT MODEL OF THE THORACOLUMBAR SPINE FOR THE INVESTIGATION OF VERTEBRAL BODY TETHERING INSTRUMENTATION

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***Abstract.*** *Anterior Vertebral Body Tethering (VBT) is an innovative fusionless growth modulating surgical technique used to treat adolescent idiopathic scoliosis (AIS), aiming to mitigate restricted motion and interference with patient growth associated with the conventional surgical treatment of scoliosis, namely posterior spinal fusion (PSF) or arthrodesis. As VBT is a relatively new approach, further research is needed to comprehensively understand its effects on the biomechanics of the scoliotic spine and to optimize its application to achieve desired outcomes while minimizing the need for revision surgery. Due to the difficulty in obtaining spine samples for experimental analyses, numerical simulation represents a viable alternative to evaluate the biomechanics of the native and instrumented spine. This study aims to calibrate and validate a robust computational model of the thoracolumbar spine (T10-S1) using experimental data from cadaveric spines. The model is developed to assess the effects of different VBT configurations (single or double cord, hybrid instrumentation, and varying tether pre-tension) on the biomechanics of the thoracolumbar spine during flexion, extension, lateral bending, and axial rotation.*

***Keywords:*** *Spine, Finite Element Method, Biomechanics, Vertebral Body Tethering*

## 1. INTRODUCTION

Idiopathic scoliosis is a three-dimensional spinal deformity with an unknown etiology, characterized by the presence of one or two abnormal curves on the coronal plane (Magee and Manske, 2021; Moramarco, 2020; Trobisch et al., 2023). It accounts for approximately 80% of all scoliosis cases and represents the most common cause of pediatric spinal deformities, primarily affecting patients between the ages of 10 and 18, known as adolescent idiopathic scoliosis (AIS) (Baker, 2020; Karavidas, 2019). Surgical treatment is often recommended for patients with scoliotic curvatures greater than 40°, and the current standard care is to perform a posterior spinal fusion (PSF) (Baker, 2020; Nicolini, 2016; Oliveira, 2022; Roach, 1999). However, long and short-term adverse effects of PSF may include decreased spinal mobility, interference in patient's growth, pseudarthrosis, acceleration of adjacent level disc degeneration, and potentially lower extremity joint problems (Baker, 2020; Oliveira, 2022; Pehlivanoglu et al., 2020; Trobisch et al., 2023).

Anterior Vertebral Body Tethering (VBT) is a fusionless growth modulating surgical technique to correct scoliotic curves while preserving the mobility of patients with AIS (Nicolini et al., 2022a; Nicolini et al., 2022c; Oliveira, 2022; Trobisch et al., 2023). Several studies have reported that VBT is a safe and effective approach to treating AIS (Baker, 2020; Hegde et al., 2021; Miyanji et al., 2020; Pehlivanoglu et al., 2020; Trobisch et al., 2019). Some authors have experimentally analyzed the effects of VBT on the biomechanics of the spine using cadaveric spine samples, yielding a better understanding of the expected mobility of patients with different configurations of VBT instrumentation (e.g., single tether, double tether and hybrid technique) under pure moment loads in flexion-extension, lateral bending and axial rotation (Lavelle et al., 2016; Nicolini et al., 2022c; Trobisch et al., 2023). Yet, there are few studies investigating the effects of VBT on the biomechanics of the spine. There is also a large revision surgery rate for VBT due to over or under-correction, and due to tether breakage (Buyuk et al., 2021; Newton et al., 2018). Thus, further understanding of how to optimize the application of VBT to ensure recovery and prevent adverse events is necessary.

Due to the limitations of cadaveric spine experimental studies (Schmidt et al., 2016), computational modeling has been a viable alternative to analyze the biomechanics of both the native and instrumented spine (Kamal et al., 2019; Nicolini et al., 2022a; Nicolini et al., 2023). Previously, a computational model of the L1-L2 segment has been calibrated and validated to numerically evaluate the impact of tether pre-tension in a single cord construct on the biomechanics of the segment for flexion-extension, lateral bending, and axial rotation (Nicolini et al., 2022a; Oliveira, 2022). Maximum tether tension during loading was also measured during the simulation, as well as coupled movements caused by the VBT implant. This information is important to predict how much tether force within VBT instrumentation is required to correct the scoliotic curve at the moment of the surgery, and whether the tension achieved in the tether is enough to cause breakage. However, VBT is usually instrumented in multiple spinal segments and therefore a larger model is required to simulate the clinical scenario. This work aimed primarily to develop a reliable and robust model of the thoracolumbar spine (T10-S1) to replicate native behavior appropriately and subsequently be used to study the effects of spine implants.

## 2. MATERIALS AND METHODS

The FE model was developed using Abaqus® (Dassault Systèmes, Waltham, MA, USA) (Figure 1). The geometric model of an adult spine was modified to represent an average adolescent spine.

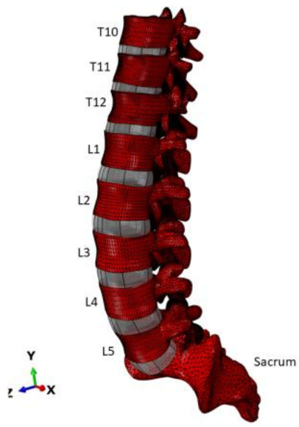


Figure 1. Geometry of the model of the native thoracolumbar spine.

The nucleus pulposus and annulus fibrosus were modeled as hyperelastic materials, using the Mooney-Rivlin model with its total strain energy function shown in Equation 1 and the model proposed by Holzapfel et al. (2000) with its total strain energy function shown in equations 2 and 3, respectively. $C_{10n}$, $C_{01n}$ and $C_{10}$ control the stiffness of the material, or its ground matrix in the case of the annulus fibrosus, while $D_n$ and D control their compressibility. Both materials were considered to be incompressible, so $D_n$ and D were set with a value of zero. The stress-strain relation of the fibers is defined by variables $k_1$ and $k_2$, while κ represents the level of dispersion of the fibers. The number N of fiber families was set to two. The term $IV_{(\alpha\alpha)}$ is an invariant determined by the square of the stretch in the fiber direction and J is the jacobian, while I and II are the first and second invariants of the right Cauchy-Green deformation tensor.

$$W_n = C_{10n}\,(I-3) + C_{01n}\,(II-3) + \frac{1}{D_n}\,(J-1)^2. \tag{1}$$

$$W = C_{10}\,(I-3) + \frac{1}{D}\left(\frac{J^2-1}{2} - \ln J\right) + \frac{k_1}{2\,k_2}\sum_{\alpha=1}^{N}\{exp[k_2\,\langle E_\alpha\rangle^2] - 1\}, \tag{2}$$

$$E_\alpha := \kappa\,(I-3) + (1-3\kappa)\left(IV_{(\alpha\alpha)} - 1\right). \tag{3}$$

The material properties of the annulus fibrous were set to vary in the circumferential and radial direction of the IVD, which was divided circumferentially into 5 symmetrical regions and radially into five more regions with homogeneous material properties. The contact of facet joints was simulated using soft contact interactions of the type nonlinear penalization (Nicolini et al., 2022b). The ligaments were modeled using bar elements with nonlinear behavior. Further details on the geometry, mesh, constitutive models, calibration of material properties of the L1-L2 intervertebral disc

(IVD), and VBT implant model utilized are described by Nicolini et al. (2022a). The mechanical properties of the L1-L2 IVD were obtained by the calibration of the model using the experimental results of a stepwise resection study of the L1-L2 segment, which allowed to isolate the mechanical behavior of the disc itself and measure the contribution of each structure (such as ligaments and facet joints) to the biomechanics of the spine in different loading directions (Beckmann et al., 2017; Beckmann et al., 2019a; Nicolini et al., 2022a; Nicolini et al., 2022c). The calibration process of the L1-L2 was performed using optimization algorithms based on the identification of the material parameters (Nicolini et al., 2022b). This task was important to obtain the material properties that approximate the numerical results to the experimental ones.

The material properties of the L1-L2 were extrapolated to the other segments. Then, a new calibration was performed by changing the material parameters of the model until obtaining a good agreement with the experimental data from the literature (Beckmann et al., 2019a; Beckmann et al., 2019b; Couvertier et al., 2017; Germaneau et al., 2016; Guan et al., 2007; Heuer et al., 2007; Jaramillo et al., 2016; Nicolini et al., 2022a; Nicolini et al., 2022c; Panjabi et al., 1994; Wilke et al., 2017; Wilke et al., 2019).

## 3. RESULTS AND DISCUSSION

### 3.1 Native spine

The calibrated properties of the thoracolumbar model are presented in Table 1. The variables $(\cdot)_c$ represent circumferential gradients and $(\cdot)_r$ represents radial gradients.

Table 1. Material properties of each IVD.

| Material Properties | T10-T11 | T11-T12 | L1-L2 | L2-L3 | L3-L4 | L4-L5 | L5-S1 |
|---|---|---|---|---|---|---|---|
| $C_{10n}$ | 0.030 | 0.030 | 0.030 | 0.015 | 0.015 | 0.015 | 0.015 |
| $C_{01n}$ | 0.19 | 0.19 | 0.19 | 0.095 | 0.095 | 0.095 | 0.095 |
| $C_{10}$ | 0.22 | 0.22 | 0.22 | 0.11 | 0.11 | 0.11 | 0.11 |
| $k_1$ | 4.5 | 4.5 | 4.5 | 5.0 | 5.0 | 5.0 | 5.0 |
| $k_2$ | 300 | 300 | 300 | 300 | 300 | 300 | 300 |
| $\kappa$ | 0.1 | 0.1 | 0.1 | 0.1 | 0.1 | 0.1 | 0.1 |
| $\beta$ | 30° | 30° | 30° | 30° | 30° | 30° | 30° |
| $k_{1c}$ | -0.05 | -0.15 | -0.05 | -0.24 | -0.24 | -0.24 | -0.24 |
| $k_{2c}$ | -0.05 | -0.15 | -0.05 | -0.24 | -0.24 | -0.24 | -0.24 |
| $k_{1r}$ | -0.15 | -0.15 | -0.15 | -0.15 | -0.15 | -0.15 | -0.15 |
| $k_{2r}$ | -0.15 | -0.15 | -0.15 | -0.15 | -0.15 | -0.15 | -0.15 |
| $\beta_c$ | 0.1 | 0.1 | 0.1 | 0.1 | 0.1 | 0.1 | 0.1 |

The numerical results for the native spine model, which includes vertebrae, IVDs, ligaments and facet joints, are shown in Figures 2 to 10 in comparison to experimental data from several studies (Beckmann et al., 2019a; Beckmann et al., 2019b; Couvertier et al., 2017; Germaneau et al., 2016; Guan et al., 2007; Heuer et al., 2007; Jaramillo et al., 2016; Nicolini et al., 2022a; Nicolini et al., 2022c; Panjabi et al., 1994; Wilke et al., 2017; Wilke et al., 2019).

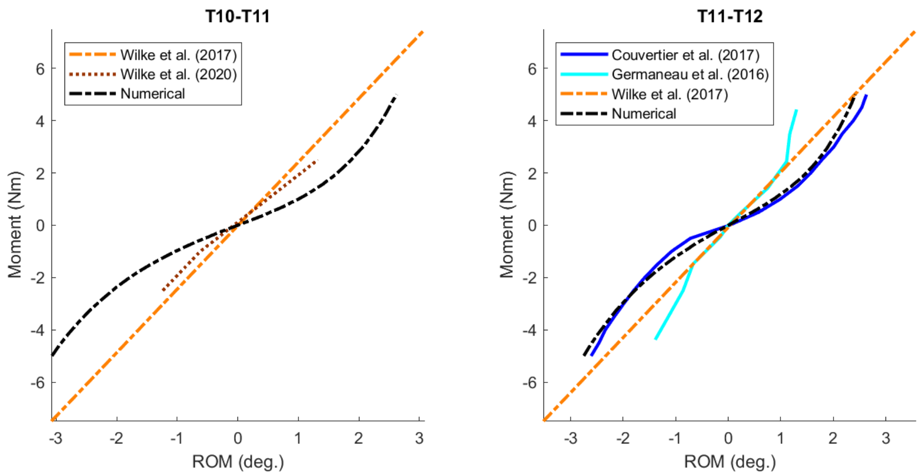


Figure 2. Numerical results compared to experimental data from the literature for the segments T10-T11 and T11-T12 of the native spine in flexion-extension.

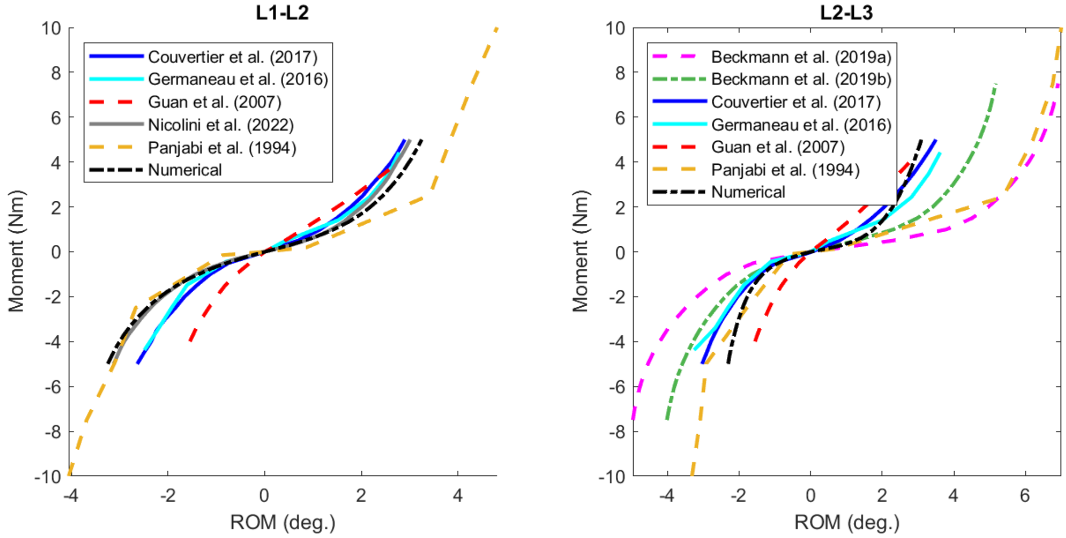


Figure 3. Numerical results compared to experimental data from the literature for the segments L1-L2 and L2-L3 of the native spine in flexion-extension.

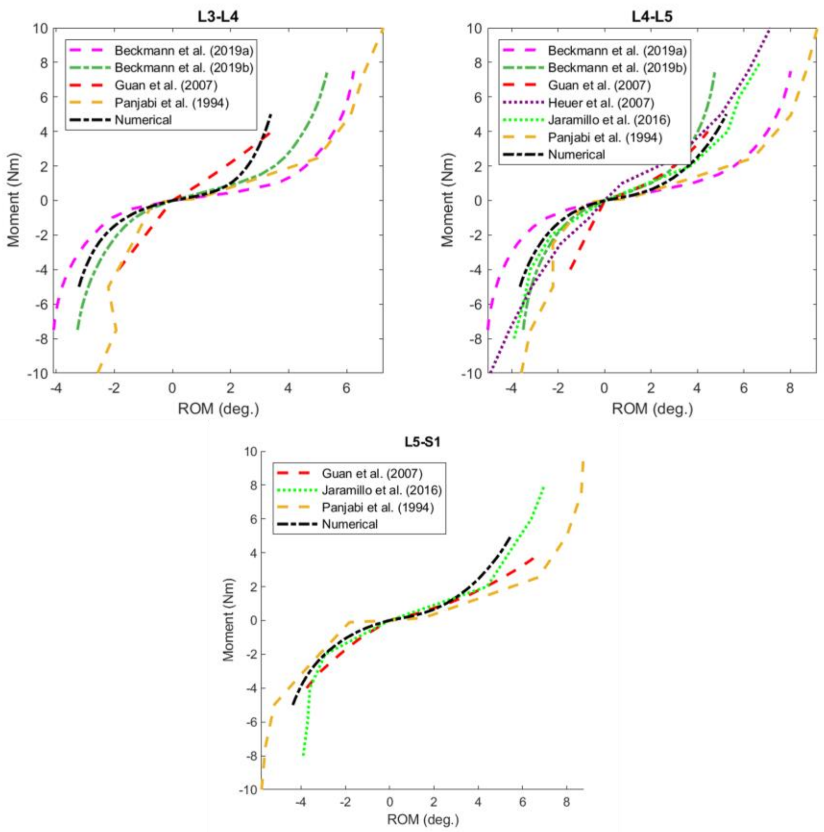


Figure 4. Numerical results compared to experimental data from the literature for the segments L3-L4, L4-L5, and L5-S1 of the native spine in flexion-extension.

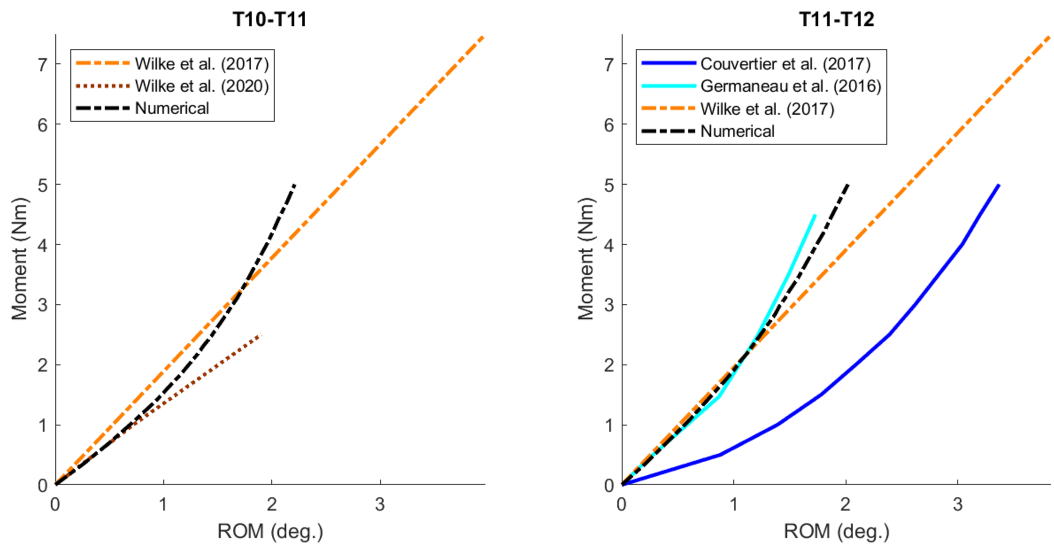


Figure 5. Numerical results compared to experimental data from the literature for the segments T10-T11 and T11-T12 of the native spine in lateral bending.

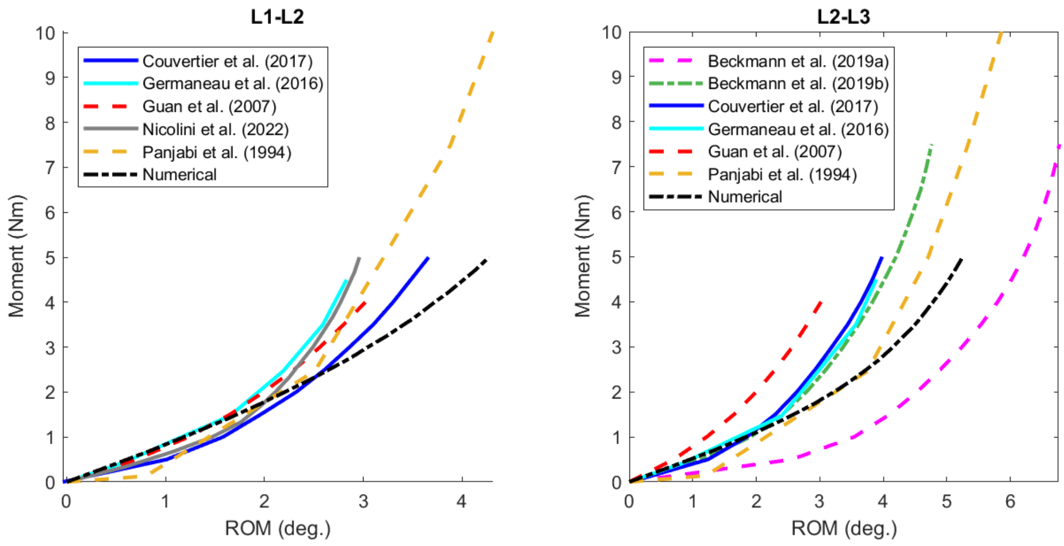


Figure 6. Numerical results compared to experimental data from the literature for the segments L1-L2 and L2-L3 of the native spine in lateral bending.

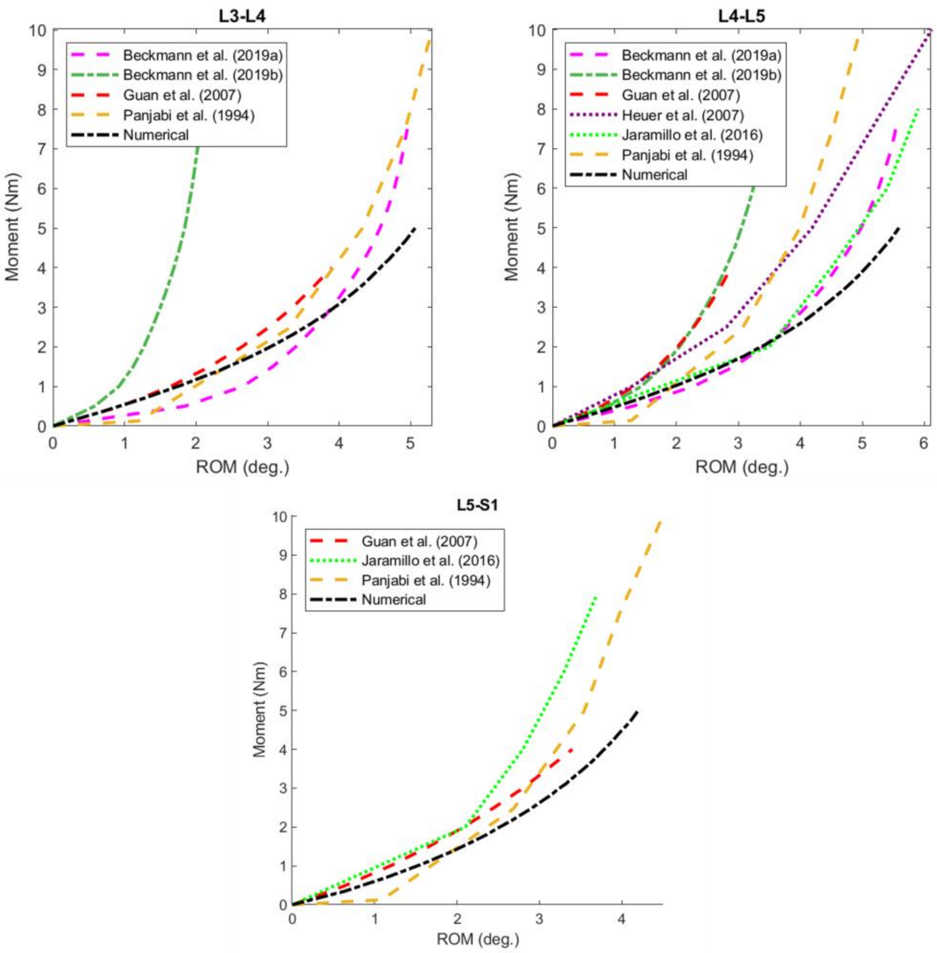


Figure 7. Numerical results compared to experimental data from the literature for the segments L3-L4, L4-L5, and L5-S1 of the native spine in lateral bending.

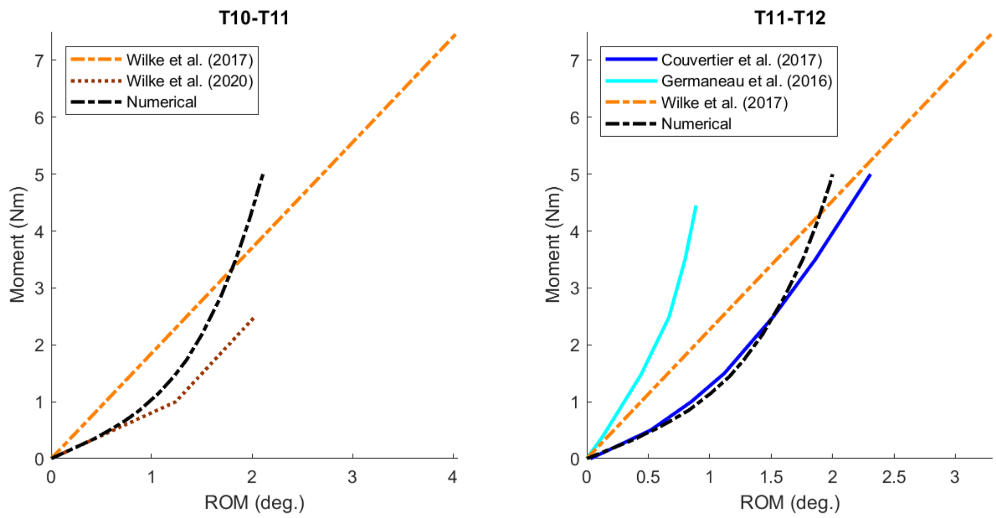


Figure 8. Numerical results compared to experimental data from the literature for the segments T10-T11 and T11-T12 of the native spine in axial rotation.

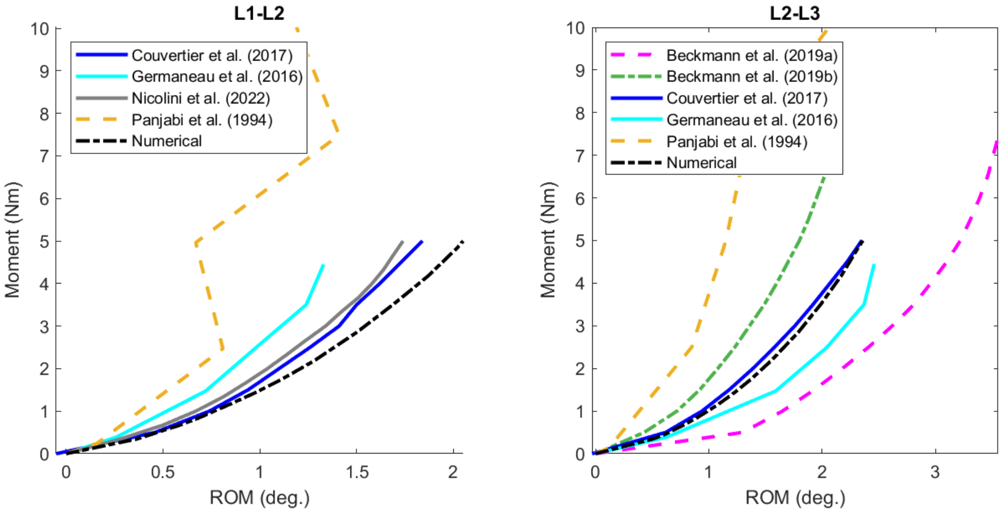


Figure 9. Numerical results compared to experimental data from the literature for the segments L1-L2 and L2-L3 of the native spine in axial rotation.

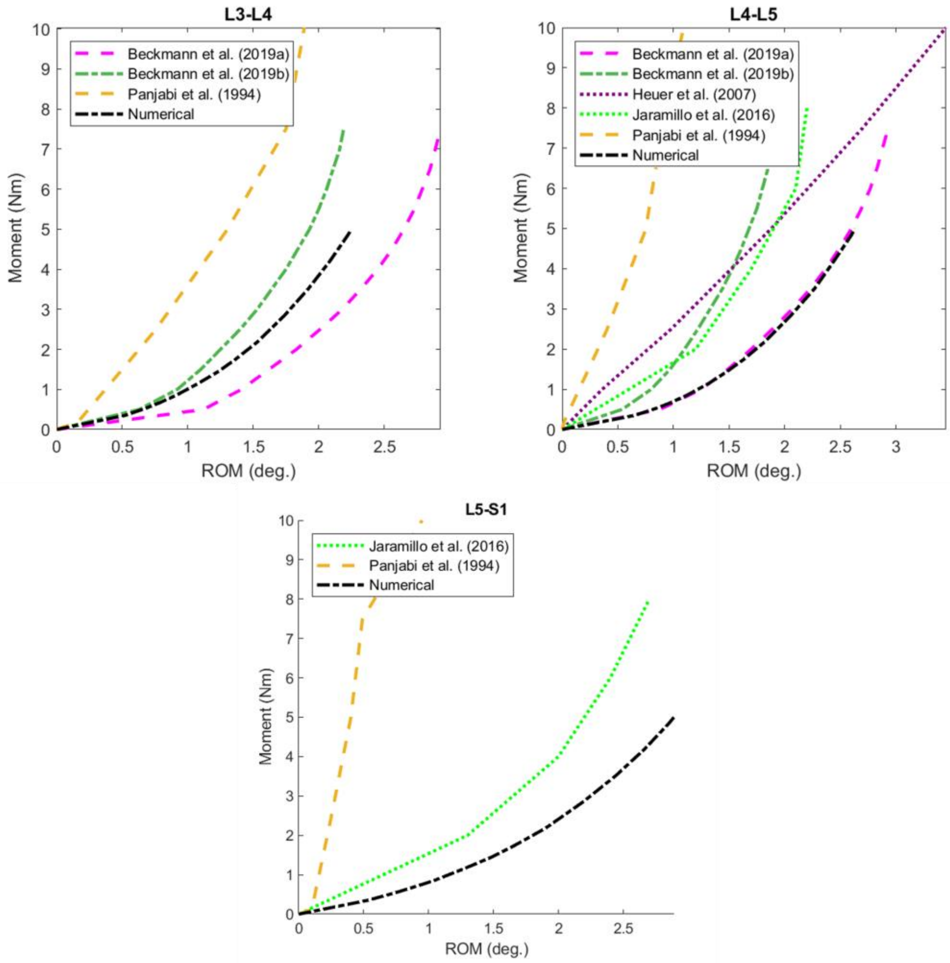


Figure 10. Numerical results compared to experimental data from the literature for the segments L3-L4, L4-L5, and L5-S1 of the native spine in axial rotation.

Figures 2 to 10 show that the model presents results that are consistent with that available from experimental data for segments T10-T11 and T11-T12. As expected, the L1-L2 segment presented numerical results close to the experimental data from Nicolini et al., since it was previously stepwise calibrated (Nicolini et al., 2022a; Nicolini et al., 2022c), while it remains close to the results from the other papers. The L2-L3 segment generated ROMs close to the experimental data from literature, while also presenting a remarkable similitude to the data from Beckmann et al. (2019b) in axial rotation and the data from Guan et al. (2007) in flexion-extension and lateral bending. The segments L3-L4 and L4-L5 presented moment-ROM curves that were contained in the range established by the literature data and presented remarkably similar behavior to the data from Beckmann et al. (2019b) in flexion-extension, axial rotation, and low angular displacements in lateral bending. The L5-S1 segment has also shown similar behavior to the reported values in the literature. Thus, the numerical results were deemed to satisfactorily represent the behavior of the thoracolumbar spine under flexion-extension, lateral bending, and axial rotation.

### 3.2 VBT instrumentation

With a reliable model of the thoracolumbar spine, VBT was inserted between the segments T10-L3. The simulations are being performed and the results will be soon reported in the literature.

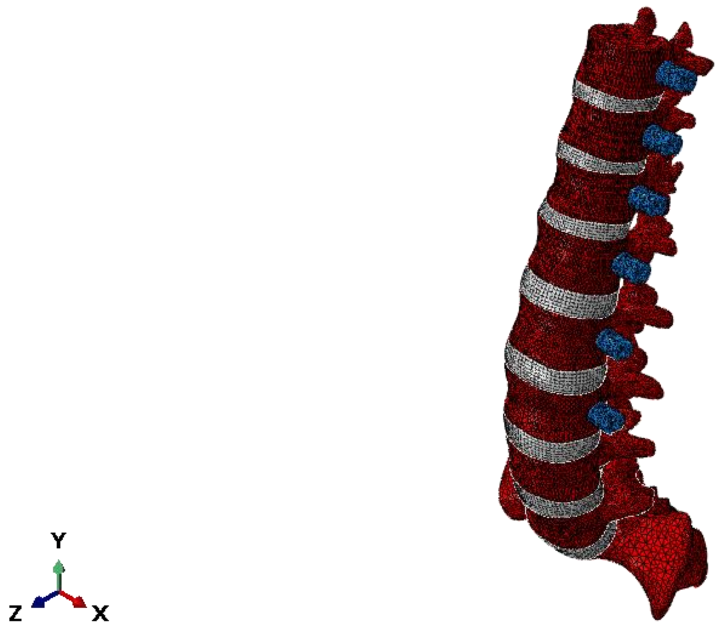


Figure 11. Model of the thoracolumbar spine (T10-S1) instrumented with single-cord Vertebral Body Tethering implant at vertebrae T10-L3.

## 4. CONCLUSION

Calibration of a spine model is a challenging task as the ligaments, facet joints, and IVD interact differently in distinct segments, with these distinctions being larger as the type of vertebra changes (lumbar, thoracic, or cervical) and as the distance between segments increases. The results obtained, however, indicate that the model reliably reproduces experimental data in flexion-extension, lateral bending, and axial rotation of the thoracolumbar spine, offering a numerical testing environment for the evaluation of different surgical techniques and implant utilization.

Our research team is currently performing numerical analyses to investigate the biomechanics of the thoracolumbar spine instrumented with VBT. The instrumented spine's ROM and the acting tension on the tether will be obtained in simulations for flexion-extension, lateral bending, and axial rotation under pure moment loading.

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