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Implementation of physiological functional spinal units in a rigid-body model of the thoracolumbar spine[☆]

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ABSTRACT

Most of the current rigid-body models of the complete thoracolumbar spine do not properly model the intervertebral joint as the highly nonlinear stiffness is not incorporated comprehensively and the effects of compressive load on stiffness is commonly being neglected. Based on published in vitro data of individual intervertebral joint flexibility, multi-level six degree-of-freedom nonlinear stiffness of functional spinal units was modelled and incorporated in a rigid-body model of the thoracolumbar spine. To estimate physiological in vivo conditions of the entire spine, stiffening effects caused by directly applied compressive loads, and contributions to mono-segmental stiffness from the rib cage as well as multi-segmental interactions in the thoracic spine were analysed and implemented. Forward dynamic simulations were performed to simulate in vitro tests that measured the load-displacement response of the spine under various loading conditions. The predicted kinematic responses of the model were in agreement with in vitro measurements, with correlations between simulated and measured segmental displacements varying between 0.66 and 0.97 ($p < 0.05$) and average deviations below 1.6°. Coupling relationships were found between lateral bending and axial rotation. Under compressive loads, the model behaved stiffer and showed a decreased range of motion: The flexion/extension response of the full thoracolumbar spine under compressive loads up to 800 N was found to strongly correlate with the literature ($r = 0.99$, $p < 0.0001$). The implementation of physiological functional spinal units with nonlinear stiffness properties into rigid-body models can enhance accuracy of biomechanical simulations, and enable detailed analysis of spinal kinematics under complex loading conditions seen in vivo.

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1. Introduction

Numerous efforts have been made in developing detailed biomechanical models of the spine based on rigid-body modelling (e.g., Abouhossein et al., 2011; Bruno et al., 2015; Han et al., 2012) or finite element method (e.g., Azari et al., 2018; Ghezlbash et al., 2016). In finite element studies, the complex structures of discs, ligaments and facet joints can be explicitly modelled in detail (Dehghan-Hamani et al., 2019; Khoddam-Khorasani et al., 2018), thereby enabling accurate kinetics analysis in various conditions. Nonetheless, the computational burden is significant (Ghezlbash et al., 2018). In contrast, musculoskeletal simulations using rigid-body models (representing the skeletal system) is another feasible option allowing spine kinematics and kinetics analysis. However,

the modelling of the intervertebral joints in the existing rigid-body models needs to be further refined, mainly in terms of the key joint stiffness properties that are currently not (properly) defined (Meng et al., 2015). The stiffness of the functional spinal unit (FSU, i.e., two adjacent vertebrae, the intervertebral joint and all tissues spanning the vertebrae) is known to passively contribute to spinal stability, kinematics, muscle forces and joint reaction forces (Cazzola et al., 2017; Izzo et al., 2013). Numerous in vitro loading-displacement tests (e.g., Heuer et al., 2007) characterized FSU stiffness properties and documented highly nonlinear, level-specific and movement direction-dependent characteristics, that strongly depend on the loading conditions.

State of the art rigid body models of the thoracolumbar spine fail to properly represent the high nonlinearity of FSU stiffness – with stiffness values usually being small for minor FSU displacements (thus inducing small passive forces) but with stiffness typically rapidly increasing with increasing displacement (Izzo et al., 2013) – thereby allowing small active muscle effort near the neutral posture while ensuring stability at the movement

[☆] The model will be available in the online OpenSim repository (simtk.org/projects/spine-stiffness).

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extremes. In finite element models, nonlinear stiffness is usually incorporated using detailed passive structures or nonlinear deformable beams (Ghezalbash et al., 2018). For rigid-body models, only a few nonlinear stiffness expressions were proposed previously and these were either only for one single FSU (Malakoutian et al., 2016; Weisse et al., 2012), for explicitly modelled ligaments (Liu et al., 2019; Liu et al., 2018) or had not been assessed comprehensively (Cholewicki et al., 1996). Whereas for the lumbar spine, some stiffness formulations are available (with being mainly linear formulations) (Christophy et al., 2012; Meng et al., 2015), for the thoracic spine, the stiffness formulations are even more limited. For instance, all thoracic FSUs in (Ignasiak et al., 2016) share similar linear stiffness properties, thereby ignoring physiologic differences across different spinal levels (Liu et al., 2018). For the thoracic FSU, interactions with the ribcage as well as other thoracic segments need to be taken into account when modelling the nonlinear FSU stiffness. Indeed, the mechanical interaction with the rib cage has a stabilizing and stiffening effect as described by (Liebsch et al., 2017; Mannen et al., 2018). Some studies (Ghezalbash et al., 2016; Ignasiak et al., 2016) approximated the contributions of the ribcage by increasing the thoracic rotational stiffness. Furthermore, some researchers (Dickey and Kerr, 2003; Kettler et al., 2000) found that the in vitro displacements in individual FSUs tend to be larger in mono-segmental test conditions, i.e. only one FSU is tested, compared to multi-segmental condition, where porcine and sheep lumbar specimen were used respectively.

Proper representation of the dependency of the FSU stiffness on compressive load magnitude is barely implemented in rigid-body models of the spine. Indeed, in vitro load-displacement tests (Patwardhan et al., 2003; Stanley et al., 2004) reported increased FSU stiffness with increasing compressive follower loads (CFLs – applied by cables that approximate the tangent to the curve of the spine and follows the displacements of vertebral bodies). In finite element simulations, the follower load is typically applied through wrapping cable elements proposed by Shirazi-Adl (2006), or other elements like the two-node thermos-isotropic truss elements as used in (Renner et al., 2007). In rigid-body models, Stokes and Gardner-Morse (2003) previously described the dependence of FSU L2-L3 stiffness on axial compressive preload using an asymptotic exponential equation, where linear stiffness

was used. To better simulate in vivo loading conditions using rigid-body models, it is necessary to extrapolate and evaluate the stiffness dependence on compressive load for the full thoracolumbar spine.

Finally, previous stiffness formulations were usually derived based from nominal or averaged experimental measures. Given the large variation among individuals, this approach can be questioned and the added value of accounting for this large variance when modelling the stiffness needs to be evaluated (Abouhossein et al., 2011).

This study therefore presents an advanced rigid-body modelling workflow for the intervertebral joints of the complete thoracolumbar spine that accounts for (1) level-specific six degree-of-freedom (DoF, three rotations and three translations) nonlinear stiffness; (2) the stiffening effect of compressive load; (3) the stabilizing effect of the rib cage and (4) multi-segment interactions in the thoracic spine and (5) potential parameter variance among individuals. Our overall hypothesis is that our extended model of intervertebral joint stiffness accurately represents thoracolumbar spinal mechanical behavior in agreement with published in vitro data.

2. Methods

An existing musculoskeletal model of the fully articulated thoracolumbar spine of a non-pathologic male adult (25 years, 175 cm, 78 kg) (Bruno et al., 2015), developed in the open-source platform OpenSim (version 3.3, Delp et al., 2007), was modified. The original model enabled 3-DoF for each spinal joint, modeled as spherical joints without explicitly modelling of intervertebral disc, ligaments and facet joints. In the current model, all 6-DoF of the intervertebral joints spanning from T1 to S1 were enabled. We used a stepwise procedure as shown in Fig. 1.

We derived an expression describing the passive force caused by FSU stiffness based on experimental data of spinal specimens (see online supplement for the details of steps 1–4):

$$F_l(x, CFL) = \int_0^x k_l(x, CFL) dx = A_l \left(e^{B_l \cdot \mu_{rib} \cdot \mu_{poly} \cdot x} - 1 \right) + r_{CFL} \cdot c_{2l} (1 - e^{c_{1l} \cdot CFL}) \int_0^x \varepsilon(x) dx \quad (1)$$

Fig. 1. Flowchart illustrating the workflow used to build the stiffness formulation of the functional spinal units (FSUs) in the thoracolumbar spine. First, we refined an existing formulation of L4-L5 FSU stiffness (Weisse et al., 2012). Second, we extrapolated this stiffness formulation to the remaining thoracolumbar FSU by tuning its parameters based on available in vitro mono-segmental load-displacement data (Panjabi et al., 1994; Wilke et al., 2017). Third, we analysed and implemented contributions from the rib cage as well as from multi-segment interactions to mono-segmental stiffness in the thoracic spine (Liebsch et al., 2017). Fourth, we statistically analysed the in vitro measurement of the FSU displacements under different compressive follower load conditions (Stokes and Gardner-Morse, 2016) and incorporated significant effects of loading on FSU stiffness. Finally, the overall integrated stiffness formulation was evaluated by simulating load-displacement tests (see Table 1) of the entire lumbar, thoracic and thoracolumbar spine.

Table 1
In vitro studies used in the forward dynamic simulation.

Spine	Study	Specimen and spine region	Mean age (years)	Sex	Compressive load	Maximum bending moment (Nm), force (N), and direction	Measured region
Lumbar	Patwardhan et al. (2003)	21, L1-S1	51.6 ± 14.6	16 male, 4 female	0, 200, 400, 800, 1200 N (along with F/E); Fix ends: L1, sacrum; Path: through L2-L5	8 in E, 6 in F	L1-S1
	Renner et al. (2007)	10, L1-S1	58 (41-73)	7 male, 3 female	0, 800 N, 1200 N (along with F/E).	8 in E, 6 in F, ±6 in LB, ±4 in AR	from L1-L2 to L5-S1
	Guan et al. (2007)	10, T12-S1	50.6 ± 13.2	-	-	±4 in F/E, LB	from L1-L2 to L5-S1
	Yamamoto et al. (1989)	10, L1-S1 (disc degenerated)	25-63	-	-	±10 in F/E, LB, AR	from L1-L2 to L5-S1
Thoracic	Liebsch et al. (2018)	8, C7-L1 (with intact rib cage)	54 ± 6	7 male, 1 female	0, 400 N; Fix ends: C7, L1; Path: through T1-T12	±5 in F/E, LB, AR	from T1-T2 to T11-T12
	Sis et al. (2016)	8, T1-T12 (with intact rib cage)	66.9 ± 4.4	4 male, 4 female	0, 200, 400 N; Fix ends: T1, T12; Path: through T3-T11	±5 in F/E, LB, AR	T1-T2, T4-T8, T8-T12
Thoracolumbar	Stanley et al. (2004)	7, T2-S1	59.9 ± 12.3	6 male, 1 female	0, 200, 400, 600, 800; Fix ends: T2, sacrum; Path: through T4-sacrum	8 in E, 6 in F	T2-S1, T2-T11, T11-L1, L1-S1
	Janevic et al. (1991)	21, L1-S1	51.6 ± 14.6	16 male, 4 female	0, 200, 400, 800, 1200 N (along with F/E); Fix ends: L1, sacrum; Path: through L2-L5	8 in E, 6 in F	L1-S1
Individual functional spinal unit	Cripton et al. (2000)	10, L1-S1	58 (41-73)	7 male, 3 female	0, 800 N, 1200 N (along with F/E).	8 in E, 6 in F, ±6 in LB, ±4 in AR	from L1-L2 to L5-S1
	Linke et al. (2003)	10, T12-S1	50.6 ± 13.2	-	-	±4 in F/E, LB	from L1-L2 to L5-S1

Note: F: Flexion; E: Extension; LB: Lateral Bending; AR: Axial Rotation.

where F_l is generalized force (forces and moments along 6-DoFs), l is spinal joint level, x is translational or rotational displacement, CFL is compressive load at level l , and k_l is stiffness. A_l, B_l, c_{1l}, c_{2l} are coefficients describing nonlinear stiffness and stiffening effect under CFL. μ_{poly} and μ_{rib_l} are stiffening factors of multi-segment interaction and rib cage respectively that are only defined for thoracic FSUs. r_{CFL} is a stiffening scaling factor of CFL effects for different FSUs. Coefficients are determined for flexion (F), extension (E), lateral rotation (LB), axial rotation (AR), anterior-posterior translation, superior-inferior translation, medial-lateral translation. ε is 1 by default, assuming that stiffness increases evenly with CFL along the entire displacement zone. As an alternative formulation, $\varepsilon(x)$ is a correction factor that accounts for the decrease of the CFL stiffening effect for increasing displacement (see online supplement for details). To account for inter-subject variability, the rotational coefficients A_l and B_l were determined on the mean value, the mean plus and minus one standard deviation (SD) of the experimental data, resulting in three sets of coefficients representative of nominal, reduced and increased spinal stiffness, hereafter referred to as 'less stiff' and 'stiffer' respectively.

We implemented this integrated FSU stiffness expression (Eq. (1)) in the thoracolumbar spine model using the ExpressionBasedBushingForce element in OpenSim, which generates forces or moments in response to the displacement between two

Fig. 2. A schematic showing the path of the compressive follower load. The load vector ideally passes the center of joint rotation of the vertebral body to prevent artefact moments or forces. For instance, for vertebra L2, the cable load passes L2 center of rotation with positive force (arrow in red) aligned with L2 axial rotation axis and negative force (arrow in green) aligned with the L3 axial rotation axis. For the top body that was fixed with the cable, only the negative cable tension was applied in the axial rotation axis of the inferior connected body. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 2

Parameters of the FSU stiffness in the thoracolumbar spine for (a) rotation, (b) translation, and (c) compressive load effect, including the coefficients A and B for the basic FSU stiffness, the coefficients c_1 and c_2 for the compressive load stiffening function and the stiffening factor r_{CFL} , multi-segmental stiffening factors μ_{poly} , as well as the rib cage stiffening factor μ_{rib} . Note for rotational coefficients A and B , three sets of parameters were given for each joint level respectively, where the bold values in the first row indicate the normal parameter and values in the second row indicate the less stiff/stiffer parameters.

(a) Rotation		Flexion				Extension				Lateral Bending				Axial Rotation					
Joint Level		A(Nm)	B(deg.-1)	μ_{rib}	r_{CFL}	A(Nm)	B(deg.-1)	μ_{rib}	r_{CFL}	A(Nm)	B(deg.-1)	μ_{rib}	r_{CFL}	A(Nm)	B(deg.-1)	μ_{rib}	r_{CFL}		
T1-T2		1.59	0.32	1.33	1.03	0.04	0.57	1.33	0.04	0.28	0.50	1.59	0.23	0.41	0.43	2.25	0.38		
		1.26/2.35	0.29/0.34			0.01/0.20	0.54/0.60			0.11/0.62	0.61/0.40			0.29/0.63	0.41/0.44				
T2-T3		3.45	0.31	1.76	2.15	0.32	0.65	1.76	0.44	0.80	0.47	1.59	0.60	0.56	0.81	3.03	1.43		
		1.56/17.28	0.30/0.23			0.11/1.60	0.60/0.65			0.73/1.91	0.31/0.77			0.46/2.17	0.54/1.11				
T3-T4		6.54	0.22	1.73	2.65	0.22	0.77	1.73	0.40	1.29	0.30	1.70	0.48	0.42	0.50	5.64	0.48		
		4.21/20.28	0.23/0.13			0.07/1.10	0.73/0.75			1.02/2.07	0.25/0.35			0.26/0.88	0.47/0.50				
T4-T5		2.26	0.53	4.45	3.00	0.31	0.75	4.45	0.54	0.99	0.39	1.78	0.55	0.82	0.41	3.93	0.71		
		0.99/11.30	0.60/0.27			0.22/0.79	0.56/1.05			0.84/1.29	0.34/0.46			0.46/1.86	0.42/0.36				
T5-T6		6.84	0.21	1.93	2.62	0.24	0.73	1.93	0.40	0.47	0.59	1.94	0.53	0.76	0.39	4.68	0.61		
		6.66/7.27	0.16/0.32			0.02/1.20	0.83/0.98			0.35/1.94	0.40/0.89			0.39/2.51	0.38/0.32				
T6-T7		2.90	0.38	1.89	2.38	0.25	0.71	1.89	0.40	0.72	0.46	1.80	0.52	1.85	0.23	2.55	0.75		
		1.53/10.47	0.43/0.21			0.01/1.25	0.83/1.01			0.26/3.60	0.45/0.38			1.10/5.01	0.23/0.18				
T7-T8		3.71	0.62	3.45	6.31	0.95	0.51	3.45	0.89	0.72	0.51	2.03	0.62	0.47	0.56	1.97	0.64		
		2.18/10.61	0.68/0.39			0.49/2.69	0.53/0.40			0.40/1.89	0.50/0.45			0.18/1.77	0.60/0.43				
T8-T9		3.44	0.38	0.99	2.82	0.68	0.69	0.99	1.03	0.90	0.56	1.34	0.92	1.20	0.44	1.82	1.15		
		0.51/17.20	0.61/0.23			0.27/3.40	0.70/0.48			0.31/3.08	0.72/0.36			1.13/2.20	0.29/0.74				
T9-T10		2.33	0.51	0.89	2.92	0.47	0.60	0.89	0.57	1.08	0.42	1.14	0.67	0.41	0.57	1.26	0.58		
		1.82/3.51	0.48/0.51			0.01/2.35	0.90/0.87			0.56/3.17	0.44/0.32			0.32/0.57	0.54/0.60				
T10-T11		3.37	0.37	1.83	2.67	0.93	0.68	1.83	1.38	1.66	0.40	0.47	0.96	0.80	0.62	1.19	1.29		
		1.58/16.85	0.42/0.17			0.58/2.52	0.58/0.70			0.46/8.30	0.54/0.20			0.55/3.77	0.44/0.77				
T11-T12(T12-L1)		3.01	0.38	–	2.47	0.44	0.66	–	0.62	1.13	0.51	–	0.98	2.17	0.46	–	2.21		
		1.66/15.05	0.39/0.18			0.02/2.20	0.98/0.69			0.56/3.63	0.55/0.36			1.52/10.85	0.34/0.41				
L1-L2		1.12	0.64	–	1.99	3.61	0.23	–	1.16	5.30	0.17	–	0.95	4.07	0.37	–	3.02		
		0.42/4.78	0.74/0.39			3.12/4.68	0.18/0.32			1.94/17.55	0.24/0.10			2.02/20.36	0.27/0.67				
L2-L3		0.62	0.54		0.85	11.40	0.13		1.93	3.72	0.18		0.69	5.99	0.30		3.44		
		0.53/0.77	0.49/0.61			11.20/11.85	0.10/0.20			2.34/8.95	0.19/0.13			0.70/29.93	0.60/0.17				
L3-L4		1.21	0.41		1.09	10.24	0.19		2.65	4.11	0.19		0.84	2.16	0.59		3.25		
		0.37/6.06	0.49/0.24			4.11/51.20	0.24/0.10			0.99/20.53	0.28/0.10			0.67/10.82	0.62/0.58				
L4-L5		1.08	0.34		0.76	4.16	0.25		1.48	4.52	0.19		0.90	4.15	0.49		4.65		
		1.15/0.95	0.25/0.52			1.76/7.99	0.26/0.29			3.99/8.37	0.13/0.31			0.70/20.75	0.57/0.79				
L5-S1		0.68	0.42		0.63	4.75	0.14		0.86	3.75	0.23		0.98	5.08	0.49		5.66		
		0.47/1.13	0.38/0.45			2.47/21.53	0.15/0.08			1.81/18.73	0.23/0.15			0.75/25.38	0.64/0.79				
μ_{poly} for thoracic spine		2.46								2.43				1.36					
(b) Translation																			
		Anterior				Posterior				Superior				Inferior				Medial-lateral	
A(N)		726				145				415				221				222	
B(mm ⁻¹)		0.45				0.65				0.993				1.81				0.738	
r_{CFL}		8.17				3.34				2.58				0.88				3.54	
(c) Compressive load effect																			
		Flexion	Extension	Lateral Bending		Axial Rotation		Anterior Translation		Posterior Translation		Superior		Inferior		Medial-lateral			
c1 (1/kN)		–0.91	–0.68	–2.43		–3.50		–1.20		–2.50		–4.60		–0.10		–2.90			
c2 (Nm/deg or N/mm)		2.87	2.75	6.39		1.78		152.96		163.96		345.05		2359.25		145.28			

bushing frames. We modified the OpenSim library of the ExpressionBasedBushingForce element to enable the implementation of CFL stiffening effect. Bushing frames had the same orientation as the body segments and were initially connected at the FSU rotation center. In the anatomic (neutral) position, each pair of bushing frames coincided. We added damping of 1Ns/mm for translation and 2.3Nm · s/rad for rotation (Abouhossein et al., 2011). Note that the damping coefficients influenced the convergence speed of our simulations but did not affect the final static poses and hence did not influence our results.

We evaluated the stiffness formulation using published in vitro data (Table 1) of the load-displacement response of lumbar, thoracic and full thoracolumbar spine as well as individual lumbar FSUs under pure moments in F/E, LB and AR (Fig. 1 – step 5). We used OpenSim's Forward Simulation tool to simulate movements of the spine and its FSUs under the experimental loading conditions. Note experimental data used here is different from the experimental data used to define the expressions for stiffness. Such cross-validation is important to avoid overfitting of the experimental data. To further assess the effects of nonlinear stiffness expressions, we included the comparison of the kinematic responses with those of two different linear stiffness expressions used in recently published rigid-body models by (Ignasiak et al., 2016; Senteler et al., 2016).

To simulate the in vitro conditions of isolated ligamentous spine, we modified the model to only keep the thoracolumbar vertebrae, pelvis and sacrum, while removing the other bones and muscles. In addition, we modified the inertial parameters to reflect vertebrae inertia instead of whole trunk inertia: the vertebral mass

was set 0.1kg for lumbar vertebrae and 0.05kg for thoracic vertebrae, and the center of mass (CoM) was set at the midpoint between the superior and inferior joint rotation centers of the vertebral body (associated sensitivity analyses are detailed in supplementary material). To simulate in vitro experiments with intact rib cage, we tripled the mass of the thoracic vertebra (0.15kg) and used the CoM position of the original model, i.e., the CoM position was kept anterior to the vertebrae representing the partitioned trunk slice.

CFL was applied by simulating the cable tension in in vitro tests (Patwardhan et al., 2003) (Fig. 2). We refer to vertebrae to which the cable was attached as 'connected bodies', and the load vector ideally passes the center of rotation of the connected body with a pair of equal but opposite loads along the local AR-axes of any two adjacent connected bodies.

All simulations started from the initial unloaded upright standing posture (all FSU displacements zero). Spine movement under the prescribed external load (constant and continuous-time bending moments with or without CFL) was simulated until the model reached static equilibrium, defined as the position where the spine posture stabilized and no further joint displacements occurred. We evaluated the range of motion (RoM) as the range in joint displacement in the final static position between each pair of simulations with opposite bending moments (see Table 1).

Goodness of the proposed FSU stiffness formulation was evaluated based on the correlation between experimental and simulated RoM using Pearson's correlation tests. Correlations were deemed significant for $p \leq 0.05$. We evaluated the slopes and intercepts of the linear relations. The intercept defines the offset of our sim-

Fig. 3. Simulated and experimental range of motion of (a) the thoracolumbar spine (T2-S1, without rib cage effect) and (b) separate lumbar spine (left) and thoracic spine (right, with rib cage) under external bending loads (without application of any compressive follower load). External loads in the simulation matched the external loads in the experimental studies used for comparison. The error bars for experimental data indicate their standard deviation, while for the current simulation data they indicate the range of the simulation results when using the corresponding 'stiffer' and 'less stiff' stiffness parameters listed in Table 2.

ulations with respect to the experimental data. Slopes below and above 1 respectively indicate an under- or overestimation of the RoM.

3. Results

Model parameters are listed in Table 2, including (1) the coefficients A and B for the basic FSU stiffness (nominal, less stiffer and stiffer), (2) the coefficients c_1 and c_2 for the compressive load stiffening function, (3) the stiffening factor r_{CFL} , (4) multi-segmental stiffening factors μ_{poly} , and (5) the rib cage stiffening factor μ_{rib} . Experimental data used for building the model was fitted with R-squared values greater than 0.97.

3.1. Kinematic response to bending loads

As shown in Fig. 3, the simulated total RoM of the full thoracolumbar spine (without rib cage), the separate lumbar and thoracic spine (with rib cage) under different loading conditions (no CFL) were all in good agreement with corresponding experiments and within one SD of in vitro data. Also, the range of the simulated range of displacement (shown as the error bars in Fig. 3) from the 'stiffer' and 'less stiff' parameter sets successfully captured the reported variance of the in vitro study. The predicted RoM variance was within one SD of the experimental data for all thoracic rota-

tions, or covered the experimental range especially for the thoracolumbar and lumbar spine.

Compared to the simulations using nonlinear stiffness expressions, the two linear stiffness expressions presented less agreement with the experiments (Fig. 3). Especially for lumbar F/E and LB, both linear expressions over- or under-estimated the experimental nominal RoM (around 16° to 45°) while the deviations for the nonlinear expression were around 3° to 14°.

The detailed segmental displacements in both the individual lumbar and thoracic model (with rib cage) were in good agreement with experiments (Fig. 4), with the average difference in RoM over FSUs in each segment being less than 1.6°. Details of the correlation and linear relations between simulations and corresponding experiments are presented in Fig. 4.

Pronounced coupling motions between LB and AR in both the thoracic and lumbar region were found. This coupling was quantified as coupling ratio (coupled motion/dominant motion). Similar coupling trends of LB/AR (Fig. 5(a)) and AR/LB (Fig. 5(b)) were observed as in (Cholewicki et al., 1996; Liebsch et al., 2018). Coupled F/E during LB or AR were negligible, with overall F/E rotation for lumbar and thoracic spine being typically less than 1°.

3.2. Kinematic responses to combined bending and compressive load

The application of compressive follower load proved to be effective: the spine remains stable presenting only negligible angular

Fig. 4. Segmental range of motion of the lumbar spine and thoracic spine (with rib cage) under external bending loads (without application of any compressive follower load). External loads in the simulation matched the external loads in the experimental studies used for comparison: lumbar spine was loaded with -8 to 6 Nm in flexion/extension, ± 6 Nm in lateral bending and ± 4 Nm in axial rotation and thoracic spine was loaded with ± 5 Nm in all three rotational directions. The error bars for experimental data indicate their standard deviation. The table shows correlation and linear relation parameters between simulations and corresponding experimental studies (listed in Table 1) in lumbar FE (Guan et al., 2007; Renner et al., 2007; Yamamoto et al., 1989), LB (Renner et al., 2007; Yamamoto et al., 1989), and lumbar AR (Yamamoto et al., 1989) as well as thoracic FE, LB, and AR (Liebsch et al., 2018; Sis et al., 2016).

Fig. 5. Coupling motions obtained during the dominant motions: (a) coupled lateral bending during (right) axial rotation; (b) coupled axial rotation during (right) lateral bending. Positive values indicate contralateral coupling while negative values indicate ipsilateral coupling. External loads in the simulation mimicked the external loads in the experimental study used for comparison and these are indicated in the brackets. The error bars for experimental data indicate their standard deviation. In (a): for the thoracic spine, the contralateral coupling relationship first increases in the first five cephalic FSUs from T1-T2 to T5-T6 then decreases or increases unevenly until reaches another local peak at end FSU T11-T12; for the lumbar spine, the ratio starts with positive values at the cephalic part, then decreases and becomes negative at the caudal part. In (b): for the thoracic spine, the absolute ratio first increases to middle thoracic region then slightly reduces; for the lumbar spine, none or small coupling ratios (less than 0.1) show with exception for FSU L4-L5 and L5-S1.

changes under large CFL. For example, the entire lumbar spine exhibited a maximum rotation less than 1° under CFL = 1200N (same as (Patwardhan et al., 2003)) and less than 3° under CFL = 5000N. The model successfully captured the decreased flexibility in response to increased CFL for individual FSUs (Fig. 6), separate entire lumbar spine, thoracic spine and full thoracolumbar spine (Fig. 7).

For individual FSU rotations (Fig. 6 (a)–(c)), the normalized rotation to zero CFL closely matched the corresponding experimental data (max CFL = 2200N): the average absolute difference was 3.4% for F/E, 3.8% for LB, 9.4% for AR. For individual FSU translations (Fig. 6(d)), both the predicted absolute and normalized RoM of lateral, anterior and posterior shear displacement showed comparable results as Janevic et al (1991). The predicted absolute values were within one SD of in vitro RoM, and the absolute differences of normalized ratios under CFL = 2200N were 1.4% for lateral shear, 6.2% for anterior shear, 0.7% for posterior shear.

For the full thoracolumbar spine (without rib cage), the normalized F/E RoM (under -8 to 6 Nm) to zero CFL (Fig. 7) were strongly correlated with experiment (Stanley et al., 2004) ($r = 0.99$, $p < 0.0001$; slope = 1.03, intercept = -3.1°).

For the lumbar spine, the simulated decrease in lumbar F/E RoM with increasing CFL significantly correlated with experiment (Patwardhan et al., 2003) ($r = 0.93$, $p = 0.0171$; slope = 0.56, intercept = 0.48°). Nevertheless, the experimentally observed decrease in RoM (Fig. 7) was slightly more significant compared to the finite element study (representing one subject) by (Shirazi-Adl, 2006). In vitro measurements of the CFL stiffening effect on lumbar LB and

AR were not found in the listed experiments in Table 1. We compared AR predictions of lumbar FSUs (CFL = 1200N, external moment = 20Nm) to (Shirazi-Adl, 2006) and found a close match: the normalized ratios of FSU RoM were 50% to 62% versus 60% to 65% for the simulations versus the finite element study.

For the thoracic spine (with rib cage), the simulated decrease in thoracic F/E and AR RoM (± 5 Nm) under CFL = 400N was also in agreement with experiments (Liebsch et al., 2018; Sis et al., 2016). The model slightly underestimated the experimentally observed decrease in thoracic LB: the decrease was 23.4% versus 38.9% in F/E, 31.2% versus 33.2% in AR, and 41.5% versus 76.8% in LB for the simulations versus experiment (Liebsch et al., 2018), and 23.8% versus 17.0% in F/E, 33.6% versus 33.2% in AR, and 48.4% versus 72.1% in LB for the simulations versus experiment (Sis et al., 2016).

4. Discussion

We implemented a level-specific, nonlinear stiffness in a full thoracolumbar spine rigid-body model. Also stiffening effects of compressive load and thoracic spine interaction were successfully integrated.

In terms of the level-specific nonlinear stiffness behavior, the simulated multi- and mono-segmental RoM in the lumbar, thoracic and thoracolumbar spine were in good agreement with literature (averaged difference below 1.6°), despite the varying experimental setups, specimen characteristics and loading conditions across the

Fig. 6. Simulated versus experimental range of motion of single lumbar joint under compressive follower loads: (a) flexion/extension, (b) lateral bending, (c) axial rotation, (d) shear displacement. External rotation moments or forces in the simulation matched the external loads in the experimental study used for comparison and these are indicated in the brackets. The error bars for experimental data indicate their standard deviation. Note normalized ratio under CFL = 4400 N were not compared as the authors stated that this magnitude exceeded the resolution of their methods. Individual FSU displacements were calculated based on averaging the displacements of all single FSUs documented in the corresponding experiments.

available studies. Wheeler et al. (2011) quantified the maximum RoM differences of single FSU between various laboratory-specific apparatuses always exceeded 1° . Thereby the observed level difference is in the order of measurement accuracy. Furthermore, the nonlinear stiffness formulation outperformed the linear stiffness expressions (Ignasiak et al., 2016; Senteler et al., 2016) in predicting the spinal kinematics.

Only a few biomechanical rigid-body models integrated the theoretical stiffness properties, let alone the stiffening effect through interaction with the rib cage (Ghezelbash et al., 2016; Ignasiak et al., 2016). Both were introduced in our study and resulted in accurate simulations comparable to experimental spinal kinematics measured in the presence (Liebsch et al., 2018; Sis et al., 2016) and absence (Stanley et al., 2004) of the rib cage. It should be noted that the rib cage effects were implicitly incorporated by stiffening FSUs, representing a simplification of the effects of the complex rib cage structure. If the rib cage and rib-vertebrae articulations are explicitly modelled, the stiffness formulation without the rib cage stiffening effect can be used. However, further validation might be necessary to evaluate the combined simulation effect.

In terms of stiffening effects from multi-segment interactions, our stiffness formulation qualitatively captured the decrease of thoracic RoM in all three dimensions in the multi-segmental situations when compared to mono-segmental FSU test. For the lum-

bar spine, since the relative in vitro RoM relationship used for modelling the FSU stiffness were from the entire lumbar spine (Panjabi et al., 1994) instead of mono-segments (see [supplementary material](#)), therefore the multiple segment interactions were incorporated implicitly. Such stiffening behavior were thought to be caused by structures like supraspinous and longitudinal ligaments that span several spinal joints but are typically cut during the mono-segmental test (Wilke et al., 2017).

Strong coupling motions were found between LB and AR, even although no coupling items were explicitly modelled. The overall coupling trend were in good agreement within vitro experiment (Cholewicki et al., 1996; Liebsch et al., 2018). This motion behavior is mainly due to the specific spine curvature (lordosis or kyphosis) (Cholewicki et al., 1996). Another contributing factor could be the adopted free joint definition with all rotational and translational DoF instead of a typically used 3DoF spherical joint. The resulting moving rotation center during the movement could also cause the secondary, coupled motions.

In terms of the stiffening effects of compressive load, we successfully simulated the CFL path in this rigid-body simulation, which approximates the spine curve, and minimize artefact moment and shear forces. Our model captured the decrease in spinal RoM under CFL as seen in in vitro experiments. The best matches were observed for the FE of the full thoracolumbar spine under CFLs up to 800 N (correlation factor $r = 0.99$; slope = 1.03),

Fig. 7. Normalized range of flexion/extension of the thoracolumbar and lumbar spine to zero compressive follower load under different compressive follower loads. External flexion/extension moments in the simulation matched the external loads in the experimental study used for comparison and these are indicated in the brackets with corresponding spinal segments. The error bars for experimental data indicate their standard deviation.

and also for individual FSU rotations and translations under CFLs up to 2200 N (averaged differences around 0.7% to 9.4%).

We observed slightly stiffer behavior in lumbar F/E compared to (Patwardhan et al., 2003; Shirazi-Adl, 2006) and slightly less stiff response in thoracic LB under CFL compared to (Liebsch et al., 2018; Sis et al., 2016). This could be caused by specimen variance e.g., age and specimen physiological conditions. In addition, the slightly stiffer estimations of lumbar F/E could be related to the experimental observation by (Linke et al., 2003; Patwardhan et al., 2003; Shirazi-Adl, 2006), who reported that the CFL stiffening effect was much more pronounced for smaller moments/rotations than larger ones. However, we did not account for the dependence of the CFL stiffening effect on FSU displacement because the experimental data (Stokes and Gardner-Morse, 2016) used to build the CFL stiffening effect was limited to only three CFL conditions ($CFL = 0, 250, 500\text{N}$) and typically smaller test RoM ($\pm 1^\circ$ F/E and AR, $\pm 1.5^\circ$ LB) than the physiological maximum RoM (Breen et al., 2019; Rozumalski et al., 2008). In addition, stiffening under CFL might be different in thoracic and lumbar spine FSUs due to differences in anatomy. Follow-up studies that test the compressive load effect on the physiological RoM of individual lumbar and thoracic FSUs are needed.

We presented a generic model of spine stiffness that does not account for geometrical factors like disc height and area or pathological conditions. Ghezalbash et al. (2016) scaled the passive properties in their finite element model by utilizing the conventional beam theory. Such scaling method is not applicable to the proposed lumped stiffness formulation. However, the formulation contains different parameter sets that account for differences in in vitro specimens, thereby capturing similar variance of in vitro load-displacement characteristics. Therefore, the model can be assumed to be less sensitive to inter-subject variance and thereby more representative for a wide range of individuals. In future work, combining the proposed generic stiffness formulation with stiffness calibration methods (e.g., optimization technique) will be developed to better describe subject-specific FSU mechanical behavior in pathological conditions.

In summary, we presented a novel model of non-linear FSU stiffness and integrated this model in an existing rigid-body model of the thoracolumbar spine. The stiffness formulations are compre-

hensive enough to capture the physiological response of spinal kinematics under various loading conditions, and are also simple and generic enough to be used for various spine modelling studies. Our model will further our insight in spinal mechanics through more physiologically-accurate biomechanical simulations.

Declaration of Competing Interest

The authors have no conflicts of interest to declare.

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Appendix A. Supplementary material

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.jbiomech.2019.109437>.

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