Xiangjie Meng

State Key Laboratory of Automotive Safety and Energy,
Department of Automotive Engineering,
Tsinghua University,
Beijing 100084, China;
Center for Advanced Orthopaedic Studies,
Beth Israel Deaconess Medical Center,
330 Brookline Avenue, RN115,
Boston, MA 02215
e-mail: mengxjchina@gmail.com

Alexander G. Bruno

Harvard-MIT Health Sciences and Technology Program,
Cambridge, MA 02139;
Center for Advanced Orthopaedic Studies,
Beth Israel Deaconess Medical Center,
330 Brookline Avenue, RN115,
Boston, MA 02215
e-mail: agbruno@mit.edu

Bo Cheng

State Key Laboratory of Automotive Safety and Energy,
Department of Automotive Engineering,
Tsinghua University,
Beijing 100084 China
e-mail: chengbo@tsinghua.edu.cn

Wenjun Wang

State Key Laboratory of Automotive Safety and Energy,
Department of Automotive Engineering,
Tsinghua University,
Beijing 100084, China
e-mail: wangxiaowenjun@tsinghua.edu.cn

Mary L. Bousein

Center for Advanced Orthopaedic Studies,
Beth Israel Deaconess Medical Center,
330 Brookline Avenue, RN115,
Boston, MA 02215;
Department of Orthopedic Surgery,
Harvard Medical School,
Boston, MA 02115
e-mail: mbousein@bidmc.harvard.edu

Dennis E. Anderson¹

Mem. ASME
Center for Advanced Orthopaedic Studies,
Beth Israel Deaconess Medical Center,
330 Brookline Avenue, RN115,
Boston, MA 02215;
Department of Orthopedic Surgery,
Harvard Medical School,
Boston, MA 02115
e-mail: danders7@bidmc.harvard.edu

Incorporating Six Degree-of-Freedom Intervertebral Joint Stiffness in a Lumbar Spine Musculoskeletal Model—Method and Performance in Flexed Postures

Intervertebral translations and rotations are likely dependent on intervertebral stiffness properties. The objective of this study was to incorporate realistic intervertebral stiffnesses in a musculoskeletal model of the lumbar spine using a novel force-dependent kinematics approach, and examine the effects on vertebral compressive loading and intervertebral motions. Predicted vertebral loading and intervertebral motions were compared to previously reported in vivo measurements. Intervertebral joint reaction forces and motions were strongly affected by flexion stiffness, as well as force-motion coupling of the intervertebral stiffness. Better understanding of intervertebral stiffness and force-motion coupling could improve musculoskeletal modeling, implant design, and surgical planning. [DOI: 10.1115/1.4031417]

1 Introduction

Multibody musculoskeletal models are widely used in biomechanics research to estimate the forces on muscles, bones, and joints, which are not easily measured in vivo [1]. Given the large

societal costs of low back problems [2,3], it is not surprising that significant efforts have been focused on developing detailed three-dimensional multibody musculoskeletal models of the lumbar spine [4–7]. The typical modeling approach assumes prior knowledge of the spine kinematics, but individual intervertebral motions are small and difficult to measure in vivo, requiring imaging approaches such as magnetic resonance imaging [8,9] or dual fluoroscopy [10–12]. Thus, studies often estimate individual intervertebral rotation as a proportion of overall spinal motion.

¹Corresponding author.

Manuscript received January 7, 2015; final manuscript received July 29, 2015; published online September 3, 2015. Assoc. Editor: Brian D. Stemper.

Fig. 1 Musculoskeletal model in OPENSIM (left), based on the lumbar spine model of Christophy et al. [6]. Each of the five lumbar intervertebral joints (L1–L2 to L5–S1, center) was implemented with 6DOFs (three translational and three rotational), with joint stiffness defined by a 6×6 stiffness matrix (right).

Mechanically, the amount a joint moves will depend both on the stiffness of the joint and the external loading applied to the joint, and this is believed to play a role in the amount of motion seen at each level of the spine [13]. Similarly, the stiffness of passive intervertebral structures (e.g., disk, facets, and ligaments) plays a critical role on the muscle forces required for equilibrium and hence affects vertebral loading [14].

Realistic inclusion of the effects of intervertebral stiffness in musculoskeletal models without detailed prior knowledge of individual intervertebral motions requires a method to determine intervertebral motions that account for intervertebral stiffness. Force-dependent kinematics has been proposed as a method to address this type of problem, wherein smaller motions (secondary kinematics) dependent on joint and tissues properties are calculated as part of the inverse dynamics solution to balance the joint reaction forces [15]. Using this approach in a musculoskeletal model of the knee greatly improves model predictions of secondary motions such as tibiofemoral and patellofemoral angles and translations, with only knee flexion used as a primary kinematic input [16]. In the spine, secondary kinematics could include individual intervertebral rotations and translations, using overall spine flexion, lateral bending, and axial rotation as primary kinematic inputs.

In order to examine the viability of the force-dependent kinematics approach in modeling of the spine, the goals of this study were to (1) implement six degree-of-freedom (DOF) intervertebral joints with intervertebral stiffness and a method to determine secondary intervertebral motions in a musculoskeletal model of the lumbar spine; (2) parametrically examine the effects of intervertebral joint stiffness magnitude and coupling on model outputs, specifically vertebral compressive loading and intervertebral motions; and (3) compare results with values measured in vivo.

2 Methods

2.1 Musculoskeletal Model. A musculoskeletal model was created in OPENSIM, an open-source musculoskeletal modeling platform [1], based on the lumbar spine model of Christophy et al. [6]. Upper extremities, lower extremities, and head–neck were added to the model (Fig. 1). Mass and inertia properties

Table 1 Joint flexion angles applied for T12–L1, L1–S1, and the hip, and IAP forces applied for the postures modeled. Individual intervertebral joint angles in L1–S1 were determined by the force-dependent kinematics algorithm to match the overall L1–S1 angle. Flexions of T12–L1 and hip were set as noted here and the joints locked during simulations.

Joint and IAP	22 deg flexion	45 deg flexion
T12–L1	1.30 deg	6.04 deg
L1–S1	15.11 deg	25.07 deg
Hip	5.95 deg	13.89 deg
IAP force	72 N (3.60 kPa)	112 N (5.60 kPa)

were adjusted to match a male with height of 170 cm and body mass of 70 kg using anthropometric data [17,18]. The model contained 238 muscle fascicles [6] implemented with the musculotendon model of Millard et al. [19]. Five lumbar intervertebral joints (L1–L2 through L5–S1) were modeled as custom joints, each with three rotational and three translational DOFs. A constant force representing intra-abdominal pressure (IAP) was applied between the pelvis and the torso (4 cm anterior to the T12 level) [20]. The force in each static posture (Table 1) was based on in vivo measurements of IAP [21], assuming a 200 cm² abdominal cross-sectional area [20] and linear variation with flexion angle [22].

2.2 Intervertebral Joint Stiffness. Intervertebral stiffnesses have been measured in a variety of studies, using both coupled and uncoupled measurement approaches. Actual intervertebral stiffness properties produce coupling between intervertebral motions, for example, with forward flexion of the joint tending to produce anterior translation as well [23,24], but most in vitro measurements of stiffness do not account for such coupling [25–30]. In the model, stiffness was applied at joint centers as 6×6 stiffness matrices using expression-based bushings (ExpressionBasedBushingForce) in OPENSIM. Coupled matrices have 11 independent non-zero stiffness parameters [23,24], resulting in the equation

$$\begin{bmatrix} M_x \\ M_y \\ M_z \\ F_x \\ F_y \\ F_z \end{bmatrix} = K_s D = \begin{bmatrix} k_{11} & k_{12} & 0 & 0 & 0 & k_{16} \\ & k_{22} & 0 & 0 & 0 & k_{26} \\ & & k_{33} & k_{34} & k_{35} & 0 \\ & & & k_{44} & 0 & 0 \\ \text{symmetry} & & & & k_{55} & 0 \\ & & & & & k_{66} \end{bmatrix} \begin{bmatrix} \theta_x \\ \theta_y \\ \theta_z \\ \delta_x \\ \delta_y \\ \delta_z \end{bmatrix} \quad (1a)$$

$$\begin{bmatrix} M_x \\ M_y \\ M_z \\ F_x \\ F_y \\ F_z \end{bmatrix} = K_s D = \begin{bmatrix} k_{11} & 0 & 0 & 0 & 0 & 0 \\ & k_{22} & 0 & 0 & 0 & 0 \\ & & k_{33} & 0 & 0 & 0 \\ & & & k_{44} & 0 & 0 \\ \text{symmetry} & & & & k_{55} & 0 \\ & & & & & k_{66} \end{bmatrix} \begin{bmatrix} \theta_x \\ \theta_y \\ \theta_z \\ \delta_x \\ \delta_y \\ \delta_z \end{bmatrix} \quad (1b)$$

where M and F represent moments and forces produced by the bushing, θ and δ are the rotations and translations of the joint, and k are elements of the stiffness matrix. Subscripts x , y , and z represent anterior-, upwards-, and right-pointing axes, respectively (Fig. 1). Uncoupled stiffness measurements only examine motion in the direction of the loading and only have nonzero parameters on the diagonal

In general, uncoupled stiffness parameters will be smaller in magnitude than the corresponding coupled parameters [24].

Coupled stiffness parameters were taken from in vitro measurements [23] of cadaveric L2–L3 and L4–L5 motion segments under 500 N preload

$$K_{L2L3} = \begin{bmatrix} 236 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & -272 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & 0 & 0 & 0 & -10.4 (\text{kN}) \\ & 734 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & 0 & 0 & 0 & 13.4 (\text{kN}) \\ & & 287 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & 9 (\text{kN}) & -5.18 (\text{kN}) & 0 \\ & & & 397 \left(\frac{\text{kN}}{\text{m}} \right) & 0 & 0 \\ \text{symmetry} & & & & 2420 \left(\frac{\text{kN}}{\text{m}} \right) & 0 \\ & & & & & 523 \left(\frac{\text{kN}}{\text{m}} \right) \end{bmatrix} \quad (2a)$$

$$K_{L4L5} = \begin{bmatrix} 377 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & -272 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & 0 & 0 & 0 & -11.6 (\text{kN}) \\ & 832 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & 0 & 0 & 0 & 13.4 (\text{kN}) \\ & & 575 \left(\frac{\text{N} \cdot \text{m}}{\text{rad}} \right) & 11.1 (\text{kN}) & -5.18 (\text{kN}) & 0 \\ & & & 473 \left(\frac{\text{kN}}{\text{m}} \right) & 0 & 0 \\ \text{symmetry} & & & & 2420 \left(\frac{\text{kN}}{\text{m}} \right) & 0 \\ & & & & & 523 \left(\frac{\text{kN}}{\text{m}} \right) \end{bmatrix} \quad (2b)$$

Because of the lack of similar coupled stiffness data measured at other vertebral levels, in this study K_{L2L3} (Eq. (2a)) was applied at levels L1–L2 and L2–L3, K_{L4L5} (Eq. (2b)) was applied at levels L4–L5 and L5–S1, and the mean of K_{L2L3} and K_{L4L5} was applied at level L3–L4. Uncoupled stiffness parameters used for each level from L1–S1, shown in Table 2, were determined from cadaveric studies reporting uncoupled measurements [29,30].

2.3 Determination of Intervertebral Angles. The primary kinematic input was overall lumbar spine flexion between L1 and

S1, with individual intervertebral flexion angles as secondary kinematics that are dependent on joint stiffness. A flowchart of the angle determination approach, implemented in MATLAB (The MathWorks, Inc., Natick, MA), is shown in Fig. 2. To account for both joint stiffness and external loading, it was assumed that the ratio of torque produced by joint stiffness to the total torque required to balance the model is the same at all levels.

$$\frac{\text{stiffness torque}}{\text{required torque}} = \frac{M_B}{M_T} = \text{ratio}_T \quad (3)$$

Table 2 Mean uncoupled stiffness parameters applied at each intervertebral joint

Parameter	Motion	L1–L2	L2–L3	L3–L4	L4–L5	L5–S1	Reference
k_{11} (N·m/rad)	Lateral bend	92	69	69	69	69	Markolf [30]
k_{22} (N·m/rad)	Axial twist	527	550	613	613	613	Markolf [30]
k_{33} (N·m/rad)	Flexion-extension	120	109	92	92	92	Markolf [30]
k_{44} (kN/m)	A–P translation	326	326	326	326	326	Lin et al. [29]
k_{55} (kN/m)	S–I translation	2038	2038	2038	2038	2038	Markolf [30]
k_{66} (kN/m)	M–L translation	476	476	476	476	476	Lin et al. [29]

The expected torque produced by each bushing can then be estimated as $M_B = \text{Ratio}_T \times M_T$, and new joint angles estimated. If the new joint angles do not add up to the desired flexion angle, the value of Ratio_T is adjusted and the process iterated until the desired amount of overall flexion is achieved.

2.4 Determination of Intervertebral Translations. It was assumed that in equilibrium the bushings would carry all of the intervertebral joint reaction forces, but correct intervertebral translations were required to produce these forces. Thus, a closed-loop algorithm was created in MATLAB to determine the intervertebral translations that equilibrate the model (Fig. 2). Simple actuators (CoordinateActuator in OPENSIM) were applied to each translational DOF in parallel with the bushing, representing the errors in bushing forces, and allowing the model to solve when bushing forces were not in equilibrium. The algorithm seeks to minimize these errors with a goal of producing actuator forces of < 0.01 N. To demonstrate the performance of the algorithm, convergence behavior was examined for the model solved for 22 deg of forward

flexion using baseline coupled stiffness (Eq. (2)). For all simulations, initial flexion angles (Table 1) were determined from the literature [12,31–33], and initial intervertebral translations set to 0, final angles and translations were calculated by the closed-loop algorithms, and the pelvis held fixed relative to the ground frame. For each cycle of simulation, static optimization was performed in OPENSIM (version 3.2) to determine muscle and actuator forces, with the objective of minimizing the sum of muscle and actuator activations squared.

2.5 Parametric Analysis of Effects of Stiffness and Coupling on Vertebral Loading in Flexion. The effects of stiffness magnitude and coupling on vertebral loading were examined for a static position of 22 deg of forward flexion. This position was chosen with the aim of producing a compressive joint reaction force of 500 N at L4–L5, which was the compressive preload applied during the measurement of the coupled stiffness matrices [23]. The flexion angle of 22 deg was determined based on variations of in vivo measurements of intervertebral disk pressure with

Fig. 2 Flowcharts of the algorithms for determining intervertebral rotations (left) and translations (right) in OPENSIM and MATLAB. When the simulation begins, the desired overall angle is entered, and initial intervertebral translations set to 0. New values for individual intervertebral angles, Ratio_T , and intervertebral translations are adjusted until the value of Ratio_T is constant at all levels and intervertebral actuator forces are < 0.01 N.

Table 3 Mean (range) of L4-L5 stiffness parameters examined in parametric analyses

Parameter	Motion	Coupled	Uncoupled
k_{33} (N·m/rad)	Flexion–extension	575 (301–849)	92 (40–287)
k_{44} (kN/m)	A-P translation	473 (317–629)	326 (217–435)
K_{55} (kN/m)	S-I translation	2420 (2104–2736)	2038 (1400–2800)

flexion [34]. Specifically, model L4–L5 compressive loading in upright stance, 332 N, corresponds to a measured disk pressure of 0.5 MPa. Estimated pressure for a load of 500 N was then $0.5 \text{ MPa} \times 500 \text{ N}/332 \text{ N} = 0.75 \text{ MPa}$, which corresponds to a lumbar flexion of 15.1 deg and the overall flexion angle of 22 deg (Table 1).

Flexion stiffness (k_{33}), A-P translational stiffness (k_{44}), and S-I translational stiffness (k_{55}) at the L4–L5 level were varied individually in both coupled and uncoupled stiffness matrices (Table 3) based on variations reported in experimental measurements. Specifically, for coupled stiffness parameters, the ranges examined were measured means ± 2 standard deviations [23], and for uncoupled parameters the ranges examined were from reported minimum to maximum values measured [29,30]. Compressive joint reaction force at L4–L5 was examined throughout the ranges of stiffnesses and compared to the expected loading of 500 N.

2.6 Parametric Analysis of Effects of Stiffness and Coupling on Intervertebral Motion in Flexion. The effects of L4–L5 stiffness magnitude and coupling on A-P and S-I intervertebral translations were examined for a single static position of 45 deg of flexion with each hand holding a 3.6 kg (8 lb) weight, allowing comparison with in vivo measurements of intervertebral motions by fluoroscopic imaging in the same posture [12]. The intervertebral flexion from L2–S1 measured by fluoroscopic imaging was 17.35 deg, with additional flexion applied at other joints to produce a total of 45 deg (Table 1). The same stiffness variations (Table 3) were used in these analyses. Flexion angles and A-P and S-I translations were determined for coupled and uncoupled stiffness matrices and compared to translations and rotations from in vivo measurements [12].

3 Results

3.1 Angle and Translation Determination Algorithm. The algorithms created succeeded in determining intervertebral angles and translations for all the values of intervertebral stiffness

Fig. 3 Convergence of intervertebral translation determination algorithm for a simulation of 22 deg of flexion, showing ActuatorForce (solid line) and bushing force (dashed line) in the axial direction. Simulation starts with zero translation input, producing large errors (ActuatorForce), but these errors rapidly converge to <0.01 N in nine cycles.

examined. The final values of Ratio_T found by the angle determination algorithm with mean coupled stiffness were 0.25 at 22 deg and 0.10 at 45 deg, while with mean uncoupled stiffness Ratio_T values were 0.197 at 22 deg and 0.068 at 45 deg. Figure 3 illustrates an example of the convergence of the translation determination algorithm. Initially, the error was high with no translation in the first cycle, and the coordinate actuators had to produce large intervertebral forces (e.g., $\text{ActuatorForce} = 820.9 \text{ N}$ in the axial direction). The translation determination algorithm converged after nine cycles, with error <0.01 N.

3.2 Effects of Stiffness and Coupling on Vertebral Loading in Flexion. Compressive joint reaction force decreased with increasing flexion stiffness (k_{33}) for both coupled and uncoupled stiffness (Fig. 4). Compressive loading at mean stiffness was

Fig. 4 Effects of flexion stiffness (a), anterior–posterior (A-P) translational stiffness (b), and superior–inferior (S-I) translational stiffness (c) on compressive joint reaction force at level L4–L5 during 22 deg of flexion. The symbols indicate mean measured stiffness values, while lines indicate realistic ranges of stiffness based on measurements. The dotted line indicates expected compressive joint reaction force of 500 N estimated based on measured disk pressure reported by Wilke et al. [34].

similar in coupled (522 N) and uncoupled (554 N) cases, and only slightly greater than the expected load of 500 N. The curves for coupled and uncoupled stiffness have similar shapes over their respective ranges of realistic stiffness values, both decreasing compressive loading by about 50 N. Compressive joint reaction force was not strongly influenced by A-P and S-I translational stiffnesses in either coupled or uncoupled models. Compressive joint reaction force decreased by about 13 N over the range of coupled A-P stiffness examined, but varied by less than 6 N over the ranges of uncoupled A-P, coupled S-I, and uncoupled S-I stiffness examined.

3.3 Effects of Stiffness and Coupling on Intervertebral Translation in Flexion. With increasing flexion stiffness, A-P translations became more anterior for uncoupled stiffness (-0.59 mm to -0.43 mm; Fig. 5(a)) but much less anterior for coupled stiffness (1.57 mm to -0.07 mm), while the magnitude of S-I translations became less inferior or compressive (uncoupled range: -0.87 mm to -0.84 mm; coupled range: -0.93 mm to -0.71 mm; Fig. 5(a)). As A-P translational stiffness increased,

A-P translations became less anterior for coupled stiffness (0.26 mm– 0.08 mm, Fig. 5(b)), but less posterior for uncoupled stiffness (-0.68 mm to -0.41 mm). The magnitude of S-I translation was not affected by A-P translational stiffness, varying less than 0.02 mm. As S-I translational stiffness increased, there was only a small effect on A-P translations with either coupled or uncoupled stiffness (uncoupled: -0.56 mm to -0.49 mm; coupled: 0.16 mm– 0.10 mm; Fig. 5(c)), while the magnitude of S-I translation decreased for both (uncoupled: -1.26 mm to -0.64 mm; coupled: -0.85 mm to -0.65 mm). A-P translations were 0.13 mm (anterior) with mean coupled stiffness but -0.51 mm (posterior) with mean uncoupled stiffness, while measured A-P translation was 1.25 ± 1.11 mm. S-I translations were -0.74 mm (inferior) with mean coupled stiffness and -0.86 mm (inferior) with mean uncoupled stiffness, while measured S-I translation was -1.1 ± 0.7 mm [12].

3.4 Effects of Stiffness and Coupling on Intervertebral Flexion Angles. With increasing flexion stiffness at L4–L5, the flexion angle at L4–L5 decreased nonlinearly for both uncoupled

Fig. 5 Effects of flexion stiffness (a), anterior–posterior (A-P) translational stiffness (b), and superior–inferior (S-I) translational stiffness (c) on A-P (left) and S-I (right) intervertebral translations at level L4–L5 during 45 deg of flexion. The symbols indicate mean measured stiffness values, while lines indicate realistic ranges of stiffness based on measurements. Dotted lines and shaded regions are the mean $\pm$ 1SD of in vivo intervertebral translations at level L4–L5 measured by Wu et al. [12].

stiffness (3.81–1.16 deg; Fig. 6(a)) and coupled stiffness (5.49–0.74 deg). Increasing A-P translational stiffness increased flexion angle for uncoupled stiffness (1.62–3.05 deg; Fig. 6(b)), but slightly decreased flexion angle for coupled stiffness (1.43–1.22 deg). Increasing S-I translational stiffness decreased flexion angle for both uncoupled stiffness (3.22–2.12 deg; Fig. 6(c)) and coupled stiffness (1.38–1.21 deg). Flexion angle was 2.12 deg with mean uncoupled stiffness but 1.28 deg with mean coupled stiffness, while measured flexion angle was 6.2 ± 4.6 deg at L4-L5 [12]. Comparing flexion between levels (Fig. 7), estimated intervertebral flexion angles were largest at L2-L3 (coupled: 9.4 deg; uncoupled: 8.3 deg), decreasing to the smallest motion at L5-S1 (coupled: 0.7 deg; uncoupled: 0.4 deg). However, previously reported measurements were on average smaller at L2-L3 (3.4 ± 5.0 deg) than at L5-S1 (6.3 ± 1.6 deg), but not significantly different between levels [12]. Flexion angle

Fig. 6 Effects of flexion stiffness (a), anterior–posterior (A-P) translational stiffness (b), and superior–inferior (S-I) translational stiffness (c) on intervertebral flexion angle at level L4-L5 during 45 deg of flexion. The symbols indicate mean measured stiffness values, while lines indicate realistic ranges of stiffness based on measurements.

Fig. 7 Intervertebral flexion angles by level estimated using coupled and uncoupled stiffness and measured by Wu et al. [12]. Error bar for measured values is + 1 SD, while error bars for model estimates show ranges found with parametric variations of L4–L5 stiffness during 45 deg of flexion.

varied at all levels with changes in L4–L5 stiffness, but with the largest effect on L4–L5 angle (0.74–5.49 deg).

4 Discussion

This study successfully implemented a force-dependent kinematics approach in a musculoskeletal model of the lumbar spine, determining individual intervertebral motions such that they depend on intervertebral stiffness. This approach allows the joint stiffness properties to make relatively similar contributions to loading at each level of the spine and prevents errors in joint kinematics or stiffness properties from producing extreme loading. This also reduces the required input kinematics to an overall motion of the spine, allowing individual intervertebral motions to arise from the model solution and provide an additional measure for model validation.

Flexion stiffness magnitude affects compressive joint reaction force similarly in realistic ranges of coupled and uncoupled stiffness values. Increasing flexion stiffness decreased compressive joint reaction forces as stiffness decreased the need for muscles (e.g., erector spinae) to balance the model. In vitro studies suggest that degenerative changes can increase intervertebral motion and reduce stiffness [35,36]. Based on the current results, this loss of stiffness could increase compressive loading produced by the musculature, potentially leading to a cycle of increasing loading causing greater degeneration. Although the uncoupled flexion stiffness values were approximately one fifth the magnitude of coupled stiffness values, the two produced similar patterns of loading. Thus, while either approach may be applicable in modeling, the actual stiffness values are not interchangeable.

Translational motions were strongly affected by flexion stiffness when coupled stiffness matrices were used. Importantly, coupled and uncoupled stiffness produced different A-P translation patterns, and coupled stiffness matched translations measured in vivo [12] better than uncoupled stiffness, demonstrating the fundamental importance of coupling on intervertebral translations. A few prior studies have implemented translational DOF in musculoskeletal models along with coupled intervertebral joint stiffness represented by a 6×6 stiffness matrix, but have examined only neutral spine postures [5,37–39] or have utilized residual force actuators to offset the forces produced by the joint stiffness [40]. In general, multibody musculoskeletal models continue to assume uncoupled motion [41], likely because of the lack of a general method to determine reasonable intervertebral translations. Importantly, incorrect translations generate large errors, thus requiring large residual forces in order to solve [40]. The current study demonstrates a method that will prevent this problem and allow intervertebral translations in future models of the spine.

The flexion angles determined were larger at higher levels than lower levels, while measurements in vivo indicate the opposite.

Possible reasons for this inconsistency include the assumption that Ratio_T is the same at all levels and the use of linear stiffness. The values found for Ratio_T suggest that coupled intervertebral stiffness was supporting about 25% of the required intervertebral torque at 22 deg of flexion, but only about 10% at 45 deg. Thus, although the torque produced by stiffness increases significantly with additional spinal flexion, the torque required to balance the spine increases more. Only linear stiffness was investigated in this study, but nonlinear stiffness would significantly increase the torque produced by stiffness at larger flexion angles. In particular, the estimated flexion angles at higher lumbar levels (e.g., about 9 deg at L2–L3) were quite large, and may in fact exceed the reasonable range of motion for this intervertebral joint [13]. Use of nonlinear stiffness properties would likely help prevent such extremes in individual joint motion, as larger motions would produce higher torques, increasing Ratio_T and redistributing angular motion to other joints. It is also possible that the assumption that Ratio_T is the same at all levels is not correct. Additional research is needed to better understand how intervertebral angles are apporportioned in vivo and develop force-dependent kinematics algorithms that match in vivo spinal motion.

There is limited data available to characterize intervertebral stiffness and particularly coupled joint stiffness, and additional in vitro measurements are needed. Finite element modeling could also be helpful in understanding the coupled and nonlinear behavior of intervertebral joints [41]. Additionally, in vivo measurements of intervertebral motions [8–12] could be combined with musculoskeletal modeling to estimate coupled intervertebral stiffness in vivo. This would allow the investigation of intervertebral stiffness under conditions of clinical importance, such as in patients with vertebral fractures, disk degeneration, facet joint arthritis, and following surgical procedures in the spine.

Surgical treatment to address back pain and spinal instability is common, with the goal of preventing abnormal intervertebral movement [42]. However, there are a variety of surgical approaches available, and the resulting joint motion may be reduced or increased relative to an intact joint [43,44]. The corresponding changes in joint stiffness could increase in vivo loading based on the results of this study. Certain disk prostheses aim to preserve normal spinal motion and may largely maintain rotational ranges of motion [45], but it is unclear if they reproduce normal translational motion or display coupled stiffness when implanted in the spine. The results of this study highlight that restoring normal intervertebral stiffness, including coupling, may be of significant importance in producing normal loading and motion, and thus is an important consideration for implant design and surgical planning.

This study has several limitations that should be noted. First, intervertebral joint stiffness was limited to linear stiffness matrices, although intervertebral joint stiffness is known to be nonlinear and affected by factors such as preload [13, 46–48]. Second, available stiffness data are limited and were not available for all vertebral levels, and thus for some levels it was estimated. Third, the parametric analyses focused on flexion motions only, and only primary (on-diagonal) stiffness terms at L4–L5 were varied. Additional work is needed to examine the effects of intervertebral stiffness in non-sagittal-plane motions. Nonetheless, this study provides a novel examination of the complex inter-relationships between intervertebral stiffness, loading, and motion.

In summary, a force-dependent kinematics approach for implementing intervertebral stiffness and determining individual intervertebral motions was applied in a model of the lumbar spine. Intervertebral joint reaction forces and motions are strongly affected by intervertebral stiffness, particularly flexion stiffness. Implementations of stiffness in musculoskeletal models of the spine must allow spinal motion to account for the stiffness properties, and stiffness properties should be carefully selected. In particular, implementations of uncoupled stiffness should not utilize coupled stiffness parameters, and vice versa. In cases of spinal instability restoration of normal intervertebral stiffness, including

coupling, may be of significant importance in producing normal loading and motion. There is a need for future work to better characterize intervertebral stiffnesses in order to understand the factors that affect intervertebral stiffness, improve model estimates of musculoskeletal loading, and inform implant design and surgical planning.

Acknowledgment

This work was supported by grants from the National Institutes of Health (NIH Grant Nos. K99AG042458, R01AR053986, and F31AG041629), National Nature Science Foundation of China (NSFC 50875151) and by doctoral visiting abroad scholarship funds from Tsinghua University.

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