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Spinal stiffness increases with axial load: another stabilizing consequence of muscle action

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Abstract

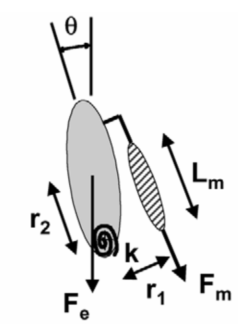
This paper addresses the role of lumbar spinal motion segment stiffness in spinal stability. The stability of the lumbar spine was modelled with loadings of 30 Nm or 60 Nm efforts about each of the three principal axes, together with the partial body weight above the lumbar spine. Two assumptions about motion segment stiffness were made: First the stiffness was represented by an 'equivalent beam' with constant stiffness properties; secondly the stiffness was updated based on the motion segment axial loading using a relationship determined experimentally from human lumbar spinal specimens tested with 0, 250 and 500 N of axial compressive preload. Two physiologically plausible muscle activation strategies were used in turn for calculating the muscle forces required for equilibrium. Stability analyses provided estimates of the minimum muscle stiffness required for stability. These critical muscle stiffness values decreased when preload effects were used in estimating spinal stiffness in all cases of loadings and muscle activation strategies, indicating that stability increased. These analytical findings emphasize that the spinal stiffness as well as muscular stiffness is important in maintaining spinal stability, and that the stiffness-increasing effect of 'preloading' should be taken into account in stability analyses.

Keywords: Spine; Stiffness; Stability; Axial preload; Muscle activation

1. Introduction

The ligamentous human spine is inherently unstable, as demonstrated by experiments showing that the entire spine can buckle with a vertical load of 20 N [19] and that the lumbar spine can buckle under a load of 88 N [5]. It is generally accepted that muscle actions stabilize the spine in vivo. Bergmark [1] provided a quantitative method to analyze the relative roles of muscle forces, muscle stiffness and elastic stiffness of the spinal motion segments in stabilizing the spine. Instability or buckling occurs when a displacement perturbation from an equilibrium position results in a force tending to increase the displacement [1], or in a net loss of the structure's potential energy [3,24] (Fig. 1).

Bergmark [1] and Crisco et al. [4] emphasized the fact that the stiffness of muscles increases with activation, and they used a linear relationship to represent this



Moment Equilibrium: $T = k\theta - F_e r_2 \mathrm{Sin}(\theta) + F_m r_1 = 0$

Stability:: $\dfrac{dT}{d\theta} > 0$

Fig. 1. Equilibrium and stability of a motion segment in one (rotational) degree of freedom, under the influence of an external force F_e and a lumped muscle force F_m. The changes in forces associated with a perturbation dθ determine whether the system is stable.

relationship for 'short-range' stiffness. Short-range stiffness is associated with rapid, small muscle length changes, and is independent of reflex actions [18,20]. Thus, qualitatively, the higher forces associated with

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heavy exertions would tend to make the trunk more unstable, but conversely the greater muscle forces required for equilibrium would increase muscle stiffness, providing a stabilizing effect. These considerations have been quantified for various lifting strategies [3] and may be altered by differing muscle activation strategies in people with back pain [21,27,28]. Muscle strategies that involve greater amounts of antagonistic muscle activation may increase stability [9], but with physiological costs of increased muscle activation and greater spinal compression [12,13,17].

Previous quantitative analyses of trunk stability have represented the spinal motion segments as elastic torsional springs [1,3,4] or as the stiffness matrices of ‘equivalent beams’ having six degrees of freedom (compression, two shears, and three rotations) [9,24]. This is in contrast to published data showing that spinal motion segments have stiffness that increases severalfold with physiological magnitudes of axial compression forces acting on them [7,10,15]. These data were obtained from tests in which motion segment stiffness or flexibility was measured with different magnitudes of axial compressive preload.

This study was designed to investigate the degree to which this stiffening effect with axial load serves to increase lumbar spinal stability. A published model of the lumbar spine and its musculature was used, with loading that simulated upper-body weight together with external forces that were pure moments (flexion, extension, lateral bending and axial rotation) applied to the thorax at the T-12 vertebra. The spine was represented as a series of beam elements representing the motion segments whose stiffness was obtained from experimental measurements of human lumbar motion segments with varying magnitudes of preload.

The objective of this study was to determine analytically the effect on trunk stability of taking into account the axial load induced alteration in spinal stiffness. An analysis in which the motion segment stiffness was held constant was compared with one in which the stiffness was updated, depending on the axial force acting on each motion segment. We tested the hypothesis that the second model would have greater stability.

2. Methods

Spinal stability analyses were performed using a quasistatic three-dimensional lumbar spine muscle model with 36 degrees of freedom (a rigid thorax and five lumbar vertebrae each having six degrees of freedom relative to the constrained sacrum). The positions of the vertebral body centers and 180 muscle attachments and the muscles’ physiologic cross-sectional areas were obtained from Stokes and Gardner-Morse [23].

The lumbar spine stiffness was obtained from direct measurement of the load–displacement behavior of four female human L2–L3 lumbar motion segments (ages 17, 21, 52 and 58) in six degrees of freedom by the method of Stokes et al. [25] using a ‘Steward platform’ (i.e. a ‘hexapod’ robot). During testing the motion segments were immersed in an isotonic saline bath cooled to approximately 4° C. Biplanar radiographs were used to establish a local axis system based on the vertebral body centers. All displacements occurred about the center of the upper vertebral body. Forces recorded by the loadcell were transformed to this same point.

Each specimen was tested with axial compressive preloads of 0 N, 250 N and 500 N following the protocol used by Gardner-Morse et al. [10,11]. The input displacements were ±0.5 mm in the AP and lateral directions, ±0.35 mm in the axial direction, ±1.5 degree in lateral bending rotation and ±1.0 degree in flexion/extension and torsional rotations. Forces were assumed to be linearly related to the displacements by a 6×6 symmetric stiffness matrix. The 21 independent coefficients of this matrix were estimated using least squares fit to experimental data [25].

Estimates of the spinal stiffness at any given axial compressive load were obtained from curvefits of the stiffness data at the three axial compressive preloads using an assumed asymptotic exponential relationship (Eq. (1)), and then intervertebral joints were represented as equivalent beams [8] having stiffness matrices whose diagonal terms were matched to these curvefits:

$$K = c_2(1-e^{c_1F}) + K_0, \qquad (1)$$

where K is a stiffness as function of axial compressive load, c_1, c_2 are coefficients determined by nonlinear least squares, F is the axial compressive load in kN, and K_0 is the stiffness with no axial compressive preload.

Since the stiffness of the motion segments was dependent on the axial compressive force, the calculation of muscle forces was performed recursively until the difference in intervertebral axial compression forces between consecutive estimates was less than 5 N.

Four different external load cases were analyzed. These were moments of 30 Nm or 60 Nm in flexion, extension, lateral bending or axial rotation at T12, representing a person making each of these four voluntary efforts in turn. An upper body weight of 340 N also acted vertically at T12 in each case.

The muscle forces required for equilibrium of the model were calculated by an optimization approach using each of two muscle activation strategies in turn. In the first strategy, the optimization cost function was equal to the sum of cubed muscle stresses. In the second strategy the cost function included both the sum of cubed muscle stresses and the sum of squared weighted intervertebral displacements, using weights that approximately equalized the contributions of the muscle stresses and the displacements [24]. In calculating the displacement component of the cost function, weights that provided for equal contributions of 1 mm of displacement and 1° of rotation were used. The sum of cubed muscle stresses cost function is considered to represent a maximum endurance strategy, and has been found to predict muscle activation patterns that are close to those observed in vivo [6,14,16,26]. The addition of the sum of squared intervertebral displacements was proposed by Stokes et al. [24] who reported that it provided estimates of muscle activations that compared favorably with observed EMG measures of muscle activation.

When using the first cost function (muscle stresses cubed) the model formulation included physiological bounds on intervertebral motion (5 mm and 5° for the sagittal plane; 2 mm and 2° for the other planes). In both cases muscle stresses were bounded in the range 0 to 460 kPa [22].

In each analysis, after the muscle forces were calculated, a critical value of the muscle stiffness parameter q was calculated in the muscle stiffness force relationship given by Bergmark [1]:

$$k = \frac{qF}{l}, \tag{2}$$

where k is the muscle stiffness, F is the muscle force, l is the muscle length, and q is a non-dimensional parameter whose value has been estimated from physiological experiments [4], and theoretical considerations based on the 'Huxley' muscle model [2]. The critical value of q was the value that made the model metastable as indicated by the smallest eigenvalue of the Hessian matrix of the trunk model's potential energy in stability analyses. The Hessian matrix gives the change in potential energy with respect to each degree of freedom of the model [3]. The spine is stable if q is greater than the critical value. In comparing the effects of updating the intervertebral stiffness based on the axial load, a decrease in the magnitude of this critical q value was interpreted as an increase in stability, and vice versa.

The amount of muscle activation associated with each simulation was calculated as the mean percent activation, where 100% activation corresponded to maximum muscle force (muscle cross-sectional area, multiplied by the upper stress bound of 460 kPa).

3. Results

The experiments with motion segments showed significant increases in stiffness in all six degrees of freedom with added preload [11]. Also, in all degrees of freedom the increases were less for the increase in preload from 250 to 500 N than for the imposition of the first 250 N (Fig. 2). This observed nonlinear relationship between preload and stiffness supported the use of an exponential curvefit in obtaining estimates of the motion segment stiffness at each preload magnitude calculated in the model analyses. The estimated values of parameters of the exponential fits (Eq. (1)) are given in Table 1.

In the analyses of stability, there was a decrease in the magnitude of the critical muscle stiffness q in all simulated conditions (external loading and cost function) when the spinal stiffness was updated for axial load (four external moment directions and two external effort magnitudes, two muscle activation strategies) (Table 2). In some cases the critical q values had small negative values, implying that the spine was stable without the need for muscle stiffness. The magnitude of the intervertebral compressive loads that were calculated ranged from 578 N to 1636 N.

In comparing the two muscle activation strategies there were different levels of stability (as quantified by critical q values) between the two cases (Table 2). The strategy that minimized intervertebral displacements as well as muscle stress cubed was associated with greater stability (lesser critical q values) in all but one case.

The mean percent muscle activation (Table 3) was greater for the second cost function that included intervertebral displacements. When the motion segment stiffening effect of preload was included in the analyses, the muscle activation was observed to decrease in all cases of the muscle stress cubed cost function, but there was minimal change in the analyses that included intervertebral displacements in the cost function.

4. Discussion

The experiments with human motion segments confirmed that the stiffness increased with preload in all six degrees of freedom. Increased stability of the analytical model was observed when this motion segment stiffening effect was taken into account for the four simulated loading directions and two magnitudes of effort. Also, this effect was observed for both of the supposed muscle activation strategies (the cost function that minimized cubed muscle stresses, and the cost function that also minimized the squared intervertebral displacements).

One of the difficulties with analyses of spinal stability is that the muscle activation pattern is not known (the 'muscle force distribution problem'). Here, we simulated two physiologically plausible strategies and found somewhat different levels of stability (as quantified by critical q values) between the two cases. In all but one case the cost function that included displacements squared predicted greater stability (lesser critical q), and this was apparently because the averaged muscle activation was greater for that cost function.

It was not necessarily expected that increased motion segment stiffness would be associated with greater stability, since the muscle forces interacted with the dis-

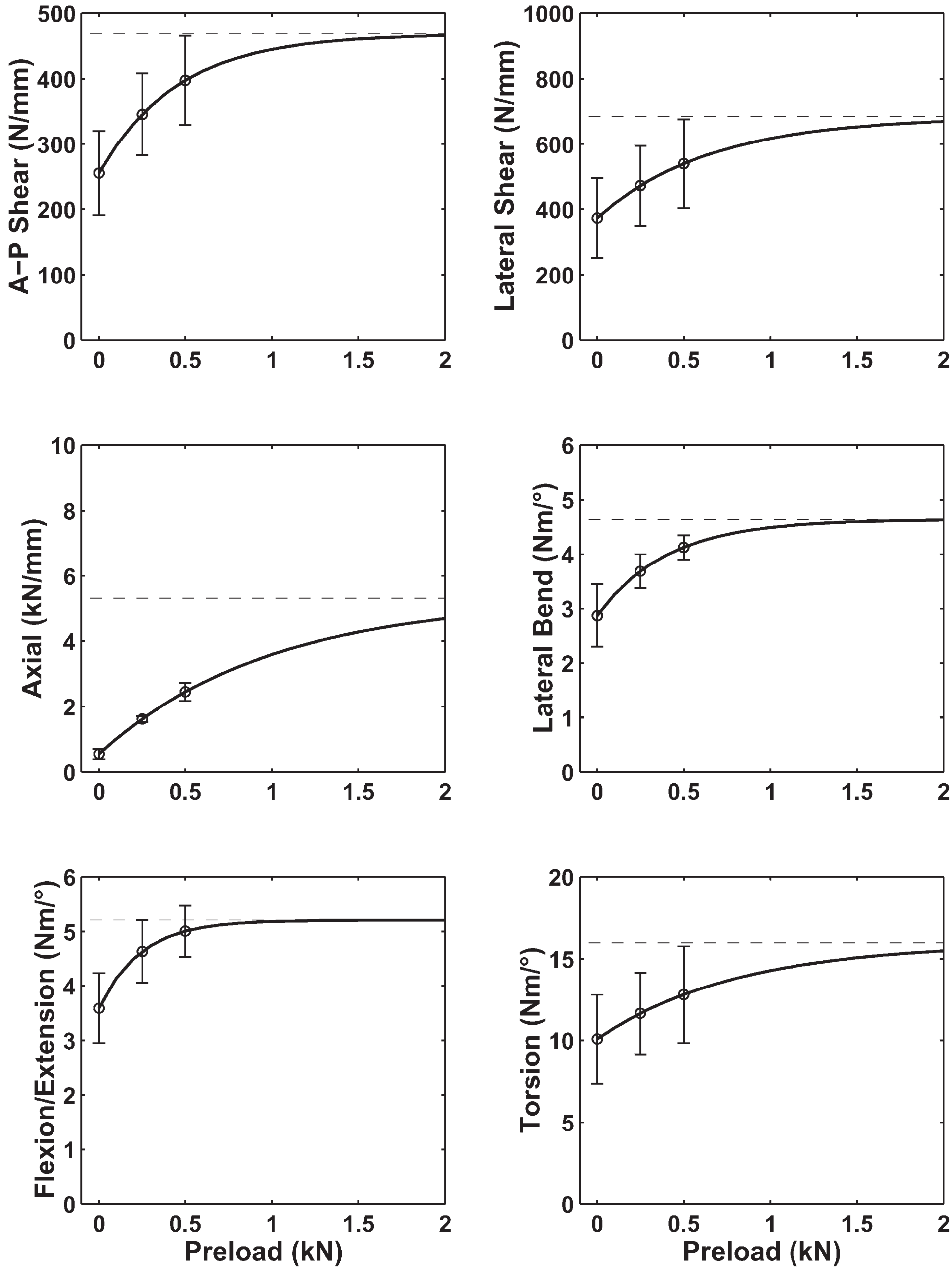

Fig. 2. Motion segment stiffness at three magnitudes of preload (0, 250 and 500 N). Each panel shows the mean and standard error for one degree of freedom, together with the exponential fit used to interpolate and extrapolate values at any specified axial load magnitude.

placement-induced (elastic) forces and torques in the motion segments that together provide equilibrium in each degree of freedom. Thus stiffer motion segments might provide forces that would lessen the amount of muscle activation, and hence the muscle stiffness. While both cost functions predicted increased stability when spinal stiffness was specified as a function of axial load, only the first cost function predicted lesser averaged muscle activation. Stability increased despite the decreased muscle activation.

Overall these simulations indicate that the dependence of the spinal motion segment stiffness on axial load, as well as the dependence of muscle stiffness on muscle activation, should be included in analyses of spinal stability.

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Table 1
Coefficients for Eq. (1) for the curvefit of increase in motion segment stiffnesses with preload

Diagonal component of stiffness matrix	c_1 (1/kN)	c_2 (N/mm or Nm/degree)	K_0 (N/mm or Nm/degree)
A–P shear	−2.1932	213.2	255.5
Lateral shear	−1.5364	310.2	373.5
Axial	−1.0214	4767.7	545.6
Lateral bend	−2.4627	1.77	2.87
Flexion/extension	−4.1294	1.62	3.59
Torsion	−1.2391	5.90	10.08

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Table 2
Critical muscle stiffness parameters (q) and increases between the two motion segment assumptions for each of two magnitudes of effort, four effort directions, and the two cost functions

Effort	Min. muscle stress cubed			Min. (muscle stress)^3 + (displacement)^2		
	Constant spine stiffness	Updated spine stiffness	Change	Constant spine stiffness	Updated spine stiffness	Change
30 Nm lateral bend	4.129	2.204	1.925	1.608	−0.281	1.889
60 Nm lateral bend	7.348	3.633	3.715	5.335	0.732	4.603
30 Nm extension	1.281	0.315	0.966	−0.497	−0.635	0.138
60 Nm extension	2.108	0.339	1.769	0.047	−0.587	0.633
30 Nm flexion	5.389	2.377	3.012	4.921	0.453	4.468
60 NmfFlexion	3.428	2.382	1.046	11.906	2.014	9.891
30 Nm torsion	1.402	0.667	0.734	−0.253	−0.416	0.163
60 Nm torsion	1.350	0.948	0.402	0.405	0.189	0.216

Table 3
Averaged percent muscle activation and increases between estimates for the two motion segment assumptions for each of two magnitudes of effort, four effort directions, and the two cost functions

Effort	Min. muscle stress cubed			Min. (muscle stress)^3 + (displacement)^2		
	Constant spine stiffness	Updated spine stiffness	Change	Constant spine stiffness	Updated spine stiffness	Change
30 Nm lateral bend	2.65	2.54	0.11	15.69	15.95	−0.26
60 Nm lateral bend	6.03	5.97	0.06	21.17	21.64	−0.47
30 Nm extension	1.90	1.84	0.06	14.03	14.85	−0.82
60 Nm extension	3.72	3.09	0.63	19.13	20.20	−1.07
30 Nm flexion	6.70	6.17	0.53	22.11	21.04	1.07
60 Nm flexion	22.57	20.41	2.16	36.06	35.68	0.38
30 Nm torsion	3.92	3.08	0.84	15.76	17.27	−1.51
60 Nm torsion	13.75	11.73	2.02	28.05	27.93	0.12

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