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 modeled 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 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...
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 several-fold 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 ±0.35 mm in the AP and lateral directions, ±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 6x6 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 curves 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 curvets:

\[ K = c_1(1-e^{-c_2F}) + K_0 \]  

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 displace-
ment component of the cost function, weights that pro-
vided 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 calcu-
lated, a critical value of the muscle stiffness parameter
q was calculated in the muscle stiffness force rela-
tionship given by Bergmark [1]:

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

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 indi-
cated by the smallest eigenvalue of the Hessian matrix
of the trunk model’s potential energy in stability analy-
ses. 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 mus-
cle force (muscle cross-sectional area, multiplied by the
upper stress bound of 460 kPa).

3. Results

The experiments with motion segments showed sig-
nificant increases in stiffness in all six degrees of free-
dom with added preload [11]. Also, in all degrees of
freedom the increases were less for the increase in pre-
load 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 para-
eters 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
element moment directions and two external effort mag-
nitudes, 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 interver-
tebral 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 (less critical $q$ values) in all but one case.
The mean percent muscle activation (Table 3) was
greater for the second cost function that included inter-
vertebral displacements. When the motion segment stiff-
ening 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 interver-
tebral displacements in the cost function.

4. Discussion

The experiments with human motion segments con-
firmed that the stiffness increased with preload in all six
degrees of freedom. Increased stability of the analytical
model was observed when this motion segment stiffen-
ing 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
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 some-
where 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 pre-
dicted 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 stab-
ility, 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.

Acknowledgements

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Table 1

<table>
<thead>
<tr>
<th>Coefficient</th>
<th>Eq. (1) for the curve of increase in motion segment stiffnesses with preload</th>
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</thead>
<tbody>
<tr>
<td>$c_1$ (1/kN)</td>
<td>$c_2$ (N/m or N/degree)</td>
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<tr>
<td>A-P shear</td>
<td>$-2.1932$</td>
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<tr>
<td>Lateral shear</td>
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<td>Torsion</td>
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Table 2

<table>
<thead>
<tr>
<th>Effort</th>
<th>Min. muscle stress cubed</th>
<th>Min. (muscle stress)$^3$ + (displacement)$^2$</th>
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</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Constant spine stiffness</td>
<td>Updated spine stiffness</td>
<td>Change</td>
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<tr>
<td>30 Nm lateral bend</td>
<td>4.129</td>
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<td>60 Nm flexion</td>
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<td>30 Nm torsion</td>
<td>1.402</td>
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<td>60 Nm torsion</td>
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<td>0.402</td>
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Table 3

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<th>Effort</th>
<th>Min. muscle stress cubed</th>
<th>Min. (muscle stress)$^3$ + (displacement)$^2$</th>
<th>Change</th>
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</thead>
<tbody>
<tr>
<td></td>
<td>Constant spine stiffness</td>
<td>Updated spine stiffness</td>
<td>Change</td>
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<td>30 Nm lateral bend</td>
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References