VISCOELASTIC MODELING OF THE LUMBAR SPINE: THE EFFECT OF PROLONGED FLEXION ON INTERNAL LOADS

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INTRODUCTION

Trunk flexion results in viscoelastic deformation of passive tissues and a consequent reduction in trunk stiffness [1]. This decrease in passive stiffness can require a compensatory increase in muscle activation [2], increasing the loads on joints and other soft tissues. In vivo methods for measuring such internal loads are invasive and costly [3], and computational modeling is a common alternative. Previous studies have provided fundamental understanding of the time-dependent behavior of trunk tissues and internal load estimation, however new approaches are needed to estimate internal loads during/following flexion exposures. Hence, the purpose of the current study was to develop a viscoelastic torso model and use it to investigate whether prolonged flexion affects internal loads during a functional task. This was done in the context of manual lifting, a common task associated with occupational low back disorders. We hypothesized that peak internal loads during a lifting task would increase following flexion exposure, and that this effect would be influenced by the angle and duration of exposure.

METHODS

A sagittally-symmetric model of the torso was developed, consisting of six deformable lumbar motion segments (i.e., T12/L1 – L5/S1 discs, ligament and facets) and one rigid superior component. Passive muscle components were modeled in the sagittal plane, with 23 local (lumbar extensor) and five global (thorax extensor) muscles. A similar wrapping mechanism as in previous work [4] was used to represent global muscle paths. Axial stiffness of each muscle, and the axial and rotational stiffnesses of each lumbar motion segment, were modeled using Standard Linear Solid (SLS) components. Previous results [5,6] were used to define material properties for each SLS component. Data from a prior study [1] were used to evaluate the viscoelastic behavior of the model.

An inverse dynamics algorithm was used to estimate muscle forces and motion segment loads during a simulated lifting task. The lifting task involved maximum trunk flexion and a 180 N load in the hands, and was performed quasi-statically over 5 sec. Kinematics of the pelvis, trunk, and lumbar motion segments and external kinetics were obtained from a previous study [4]. These were entered into the viscoelastic model to estimate the reactive moment at each level of the spine. Using these moments, a separate algorithm estimated active forces with an objective of minimizing sum of cubed muscle stresses [4]. Estimated internal loads at L4/L5 were converted into intradiscal pressures following established relationships [7], and evaluated in comparison to reported in vivo values in several flexed postures [3].

Effects of flexion angle and exposure duration on internal lumbar loads were assessed in 12 combinations of three flexion durations (2, 4, and 16 minutes) and four trunk flexion angles (40, 60, 80, and 100% of the flexion-relaxation (FR) angle). Two lifting tasks were simulated, before and after each exposure. Parameters associated with motion segment failure were estimated during the lifting tasks: peak internal load, peak axial stiffness, and absorbed energy at L5/S1. Changes in these parameters were then compared between the two lifting tasks (after vs. before flexion exposure).

RESULTS AND DISCUSSION

Good agreement between model-based and experimental results was evident both for internal load estimation and viscoelastic behavior (Fig. 1). Estimated internal loads, axial stiffness, and
absorbed energy all increased following flexion exposures, and these effects were magnified by increasing flexion angle and duration (Fig. 2). Comparable effects were found after both 4 and 16 minutes of exposure, indicating that most of the moment drop (load-relaxation) occurred within 4 minutes of exposure. A similar duration has been reported using direct measures in response to prolonged flexion \[1,5\]. For the extreme exposure condition (i.e., 16 minutes and 100% FR), passive moment reduction was \(~35\%\) (Fig. 1b), and this caused \(~8.9\%\) increase in internal loads. More detailed investigation of load partitioning among passive components indicated that \(~0.6\) Nm of the total moment drop was caused by load-relaxation in passive muscle components, which is only \(~3\%\) of the total moment drop. As such, for angles smaller than FR, the majority of the moment drop resulted from viscoelastic responses of spinal motion segments that led to additional muscle activity.

**Figure 1:** (a): Peak intradiscal pressure at L4/L5 in performing quasi-static tasks from Wilke et al. (2001) and the current model with different configurations. Effects of co-activation were simulated by activating the abdominal muscles at 0 or 1.7% of maximum forces, and with/without wrapping. (b): Predicted load-relaxation responses of the whole trunk from the model and earlier data for 100% FR exposures \[1\].

There was a limitation in the current model due to inadequate experimental data regarding the nonlinear viscoelastic behavior of passive tissues, which necessitated some assumptions for defining material properties for different muscle groups and different loading magnitudes. To overcome these potential sources of error, additional experimental results are required both for muscles and spinal motion segments at several loading magnitudes.

**Figure 2:** Percentage changes (after vs. before flexion exposure) in L5/S1 peak internal load, peak axial stiffness, and absorbed energy during simulated lifting tasks. Prior to exposure, respective values were 3201N, 708652N/m, and 10.2J.

**CONCLUSIONS**

Epidemiological evidence indicates an increased risk of low back disorders due to working in environments that require prolonged trunk flexion in combination with repetitive lifting. Current results demonstrated an increased contribution of active components required to complete a lifting task following load-relaxation exposures, consistent with the noted epidemiological evidence.

**REFERENCES**


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