THE EFFECT OF MUSCLE LOADING ON INTERNAL MECHANICAL PARAMETERS OF THE LUMBAR SPINE: A FINITE ELEMENT STUDY

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INTRODUCTION

The human spine experiences complex loading in vivo; however, simplifications to these loading conditions are commonly made in computational and experimental protocols. Pure moments are often used in cadaveric preparations to replicate in vivo loading conditions, and previous studies have shown this method adequately predicts range of motion behavior (1, 2). It is unclear what effect pure moment loading has on the tissue-level internal mechanical parameters such as stresses in the annulus fibrosus and facet contact parameters. Recent advances in musculoskeletal modeling have elucidated previously unknown quantities of the musculature recruitment patterns such as times, forces, and directions. The advancements are especially relevant in cases of surgical intervention because the spinal musculature has been reported to play a critical role in providing additional stability to the spine when defects such as discectomy and nucleotomy are involved (2). Thus, the aim of the study was to determine the importance of computational loading conditions on the resultant global ranges of motion, as well as the tissue-level predictions of annulus fibrosus stresses, and facet contact pressures, forces, and areas.

METHODS

Lumbar muscle forces were applied to L1 through L4 on a previously validated nonlinear finite element lumbar spine model (3). Muscle forces were determined using the AnyBody Modeling System (4) (AnyBody Technology A/S, Aalborg, Denmark) for the postures of 40° forward flexion (Sacrum-L1), maximum lateral bend (30° Sacrum-L1), and maximum axial rotation (20° Sacrum-L1). A total of 135 discrete forces were implemented on the finite element lumbar spine model. The force of each muscle fascicle was applied to a node set on the lumbar model via an attachment node and coupling constraint. Eight fascicles of longissimus thoracis pars lumborum, eight fascicles of iliocostalis lumborum pars lumborum, eleven fascicles of longissimus thoracis pars thoracis, and eight fascicles of iliocostalis lumborum pars thoracis were attached to form the erector spinae. Thirty total fascicles of the multifidus, four fascicles of quadratus lumborum, and eight fascicles of psoas major were attached to the spinal column. The transversus abdominal muscle and semispinalis muscles were also represented on the spine model. In vivo range of motion and intervertebral disc pressures were compared to those obtained by the finite element model in flexion, lateral bending, and axial rotation.

In order to determine the effect of muscle loading on internal mechanical parameters versus a pure moment, the previously determined muscle forces were applied to a healthy spine, a spine in which a full nucleotomy was performed on the L3-L4 disc, and a spine in which an artificial disc was implanted at the L3-L4 level. The moment magnitude was increased in all three principal directions until equal rotation was achieved between the healthy moment and muscle-driven models. Range of motion, facet joint mean contact pressures, total contact forces, total contact areas, and von Mises stresses of the annulus fibrosus were compared between the muscle-loaded and moment-loaded models.

RESULTS

Total rotation in flexion for the healthy spine with muscle forces and pure moment application was 29°. Rotation increased for both nucleotomy models to 32°. Implantation of an artificial disc decreased total rotation in flexion for both loading scenarios. Total rotation varied by a maximum of 4° for all models in lateral bending. Axial rotation range of motion results were within 2.5° for all models with the exception of the muscle-loaded nucleotomy case which decreased by 3°, likely because of increased facet contact (Figure 1).
Negligible facet contact occurred in flexion for the healthy and nucleotomy models for the two loading scenarios. Mean facet contact pressures and total contact areas for lateral bending and axial rotation are presented in Figure 2. Nucleus removal had little impact on total facet joint contact force for the muscle and pure moment loading modes in lateral bending, while the total facet contact forces increased from 2 N and 9 N in the healthy models to 240 N and 70 N in the artificial disc models under muscle and moment loading respectively. Total facet contact force for axial rotation was unchanged for all loading modes with the exception of the muscle-loaded artificial disc model where total force doubled from that of the healthy model.

Average von Mises stresses of the annulus fibrosus for flexion and lateral bending are shown in Figure 3. Annulus stresses remained constant for the healthy and nucleotomy cases for both muscle and moment models in axial rotation. Stresses increased among all levels with the implantation of an artificial disc for the muscle model, while adjacent level stresses decreased for the artificial disc model under moment loading.

DISCUSSION

Finite element models are advantageous for predicting internal mechanical parameters that cannot be easily measured experimentally. Kinematic predictions are common validation criteria; however, it is questionable whether they alone are sufficient for ensuring the accuracy of other resultant model predictions. Limitations in the application of anatomical in vivo loads to finite element models also call into question the physiologic relevance of resultant predictions. Little variation in spinal kinetics was experienced for the healthy and nucleotomy models under muscle and moment loading.

Muscle loading created the most uniform stress distribution among spinal levels in flexion and aided in spinal stabilization when a nucleotomy was performed, especially at the operated level (Figure 3). The lower stability of the spine created by the insertion of the artificial disc was accommodated by increased facet joint contact and altered intervertebral disc loading at the operated and adjacent levels. These data support the concept that pure moment application transmits uniform loading throughout the spinal column of a healthy spine, but the distributed loading pattern of the musculature may aid in providing additional support when the spinal anatomy is substantially altered.

REFERENCES