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Technical note

Effect of optimization criterion on spinal force estimates during asymmetric lifting

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Abstract

Optimization-based muscle force prediction models of the lumbar region are used in research and ergonomic practice, usually as a subroutine of a job analysis software package. Various optimization criteria have been put forward for use in rationally selecting a set of muscle forces to satisfy moment equilibrium, including the sum of cubed muscle contraction intensities and a double linear programming procedure for minimizing the spinal compression force and maximum muscle contraction intensity. A laboratory study was conducted to determine whether these two model formulations produce significantly different estimates of spinal forces for a dynamic asymmetric lift. Although statistically significant differences were found between the predictions of the two models, the difference in peak spinal compression force was only 1.1%. © 2000 Elsevier Science Ltd. All rights reserved.

Keywords: Lumbar spine; Optimization; Lifting; Muscle forces

1. Introduction

Muscle force prediction is a critical element of estimating stresses on lumbar spine. Highly anatomically detailed models of the lumbar spine have been developed (Stokes and Gardner-Morse, 1995; Cholewicki and McGill, 1996). Due to the large data requirements and model complexity, these models are not widely used in the study of manual materials handling and occupational biomechanics. Commercial computer software applications for ergonomics job analysis, which are widely available, use single equivalent models for two-dimensional lifting models and single cutting plane models with multiple muscles for three-dimensional analyses.

Several performance criteria have been proposed for the multiple muscle optimization models, including minimizing the spinal compression force while using the lowest possible muscle contraction intensity limits (Bean et al., 1988) and minimizing the sum of the cubed muscle intensities (Crowninshield and Brand, 1981). Muscle contraction intensity is the force generated by the muscle divided by the muscle's physiological cross-sectional area. The model of Bean et al. (1988) is implemented in commercial ergonomics software because it can be quickly solved on a microcomputer using a standard linear programming algorithm instead of a nonlinear method.

Thus, the purpose of this study was to assess whether choice of optimization criterion affects spinal compression force estimates in an asymmetric lifting task commonly found in the work place. If the two models provide very similar estimates of spinal load for this task, it is reasonable to continue using the computationally simpler implementation of the Bean et al. (1988) model formulation for ergonomic job analyses.

2. Materials and methods

Twenty-two subjects (8 male; 14 female) participated in the study. Median age was 30 (range 19–40) and median height was 173 cm (range 162–188). Median mass was 73 kg (range 50–106). Informed consent was obtained from all subjects. The protocol was approved by the Human Subjects Review Board of the National Institute for Occupational Safety and Health.

Each test subject lifted a 9.4 kg box from a location 3 cm in front of the toes in the mid-sagittal plane to a 79 cm tall stand located 60° to the right (Fig. 1). Each

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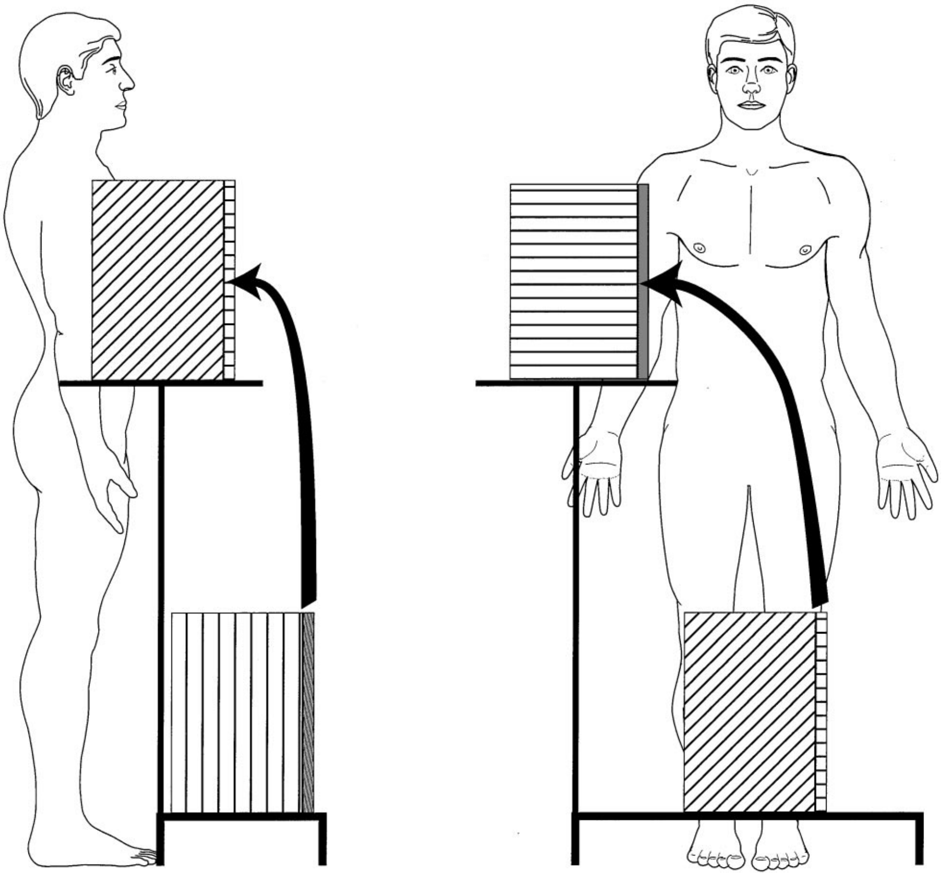

Fig. 1. Front and side views of experimental setup. A 9.4 kg frame box was lifted from a position in front of the body to a stand 60° to the right of the subject.

subject was instructed that the lift could be performed freestyle.

Each subject stood on two force plates (Model 4060-08, Bertec Corporation, Worthington, OH, USA). Kinematic parameters were measured using six ProReflex motion capture units (Qualisys, Glastonbury, CT, USA) arranged in a semicircle around the force plates. Data were sampled at 120 Hz. Kinematic data were collected on a Macintosh Power PC computer; force plate data were collected using a National Instruments AT-MIO-16E-10 DAQ card on a separate microcomputer using a LabView interface. Data collection was synchronized using a TTL output signal from the ProReflex system. The small time delay between the two systems was accounted for in the data analysis. Spherical reflective markers were placed bilaterally at the lateral malleolus, fifth metatarsal head, heel, tibial tuberosity, lateral femoral condyle, greater trochanter, and anterior superior iliac spine. Three markers were attached to a wooden base that was taped over the sacrum. This "triad" of markers was used to quantify pelvis location and orientation.

Intersegmental moments and forces were estimated from the force plate and motion data using an adaptation of the bottom-up inverse dynamic model developed in Vaughn et al. (1992). The model incorporates rigid segments corresponding to the foot, shank, and thigh. The model was adapted for use in lifting by adding a pelvis segment, and the mass and moment of inertia data for the pelvis were taken from Zatsiorsky and Seluyanov (1981). The location, velocity and acceleration of the pelvis were estimated from the triad of markers over the sacrum. The location of L3/L4 was estimated from the triad markers using the data of Tracy et al. (1989) that estimates the intervertebral center from surface locations.

L3/L4 kinetics were used as input to the double linear optimization model of Bean et al. (1988), which will be referred to here as the minimum intensity-compression (MIC) model. The same moments were analyzed using the sum of cubed muscle intensities (SCI) model. Five bilateral muscle pairs were included in the model: erector spinae, rectus abdominis, internal oblique, external oblique, and latissimus dorsi. The muscle moment arms, lines of action, and cross-sectional areas were taken directly from Hughes et al. (1994). Muscle forces were predicted for each subject's lift, and spinal compression and shear forces were estimated from the muscle forces and net intersegmental forces estimated by inverse dynamics using MATLAB version 5.2. The Student's *t*-test was used to test for differences in peak and average forces produced by the two models.

3. Results

Although there were statistically significant differences in spinal loads predicted by the two models, the differences were very small in magnitude. The mean peak compression force was predicted to be 64 N greater by the MIC model, and the average compression force was predicted to be 49 N greater (Tables 1 and 2). These differences represented only 1.1 and 1.5% of peak and average spinal compression, respectively. Increases in erector spinae forces predicted by the MIC model were nearly compensated for by increases in the oblique and latissimus dorsi muscles. Statistically significant differences were found between model predictions for the peak left erector spinae, right erector spinae, left internal oblique, right internal oblique, left latissimus dorsi, and right latissimus dorsi muscles (Table 1). Statistically significant differences in average force were also found for the same

Table 1
Peak predicted forces, using the SCI and MIC models

Quantity	SCI model	MIC model	SCI − MIC
Left erector spinae (N)[c]	2217 (806)	2014 (714)	203 (103)
Right erector spinae (N)[c]	2140 (740)	2014 (714)	125 (56)
Left rectus abdominis (N)	16 (22)	30 (51)	− 14 (39)
Right rectus abdominis (N)	16 (22)	19 (30)	− 3 (19)
Left internal oblique (N)[c]	217 (179)	287 (215)	70 (45)
Right internal oblique (N)[a]	96 (73)	140 (99)	− 44 (74)
Left external oblique (N)	272 (131)	274 (126)	− 2 (17)
Right external oblique (N)	103 (94)	101 (89)	2 (15)
Left latissimus dorsi (N)[c]	130 (49)	315 (112)	− 185 (63)
Right latissimus dorsi (N)[c]	122 (40)	315 (112)	− 193 (72)
Compression (N)[c]	5696 (1951)	5760 (1978)	− 64 (37)
AP shear (N)[c]	− 498 (388)	− 407 (409)	− 91 (41)
Lateral shear (N)[b]	207 (122)	173 (112)	34 (43)

Note: Entries are means across subjects (and the standard deviation). Note a positive AP shear value represents posteriorly directed force acting on the superior vertebra by the inferior vertebra; positive lateral shear represents force directed to the right.
[a] $p < 0.05$.
[b] $p < 0.01$.
[c] $p < 0.001$.

Table 2
Average predicted forces during the lift using the SCI and MIC models

Quantity	SCI model	MIC model	SCI − MIC
Left erector spinae (N)[c]	1163 (473)	1055 (416)	108 (63)
Right erector spinae (N)[c]	1112 (436)	1055 (418)	57 (35)
Left rectus abdominis (N)	0 (0)	3 (9)	− 3 (9)
Right rectus abdominis (N)	0 (0)	0 (0)	0 (0)
Left internal oblique (N)[c]	77 (96)	102 (120)	− 25 (24)
Right internal oblique (N)[c]	22 (23)	29 (30)	− 7 (7)
Left external oblique (N)[a]	106 (71)	108 (70)	− 2 (5)
Right external oblique (N)	9 (17)	9 (16)	0 (3)
Left latissimus dorsi (N)[c]	68 (28)	165 (65)	− 97 (37)
Right latissimus dorsi (N)[c]	60 (27)	165 (66)	− 105 (43)
Compression (N)[c]	3517 (1187)	3566 (1198)	− 49 (25)
AP shear (N)[b]	− 166 (351)	− 100 (367)	− 66 (28)
Lateral shear (N)[c]	87 (101)	70 (83)	17 (25)

Note: Entries are means across subjects (and standard deviation). Note a positive AP shear value represents posteriorly directed force acting on the superior vertebra by the inferior vertebra; positive lateral shear represents force directed to the right.
[a] $p < 0.05$.
[b] $p < 0.01$.
[c] $p < 0.001$.

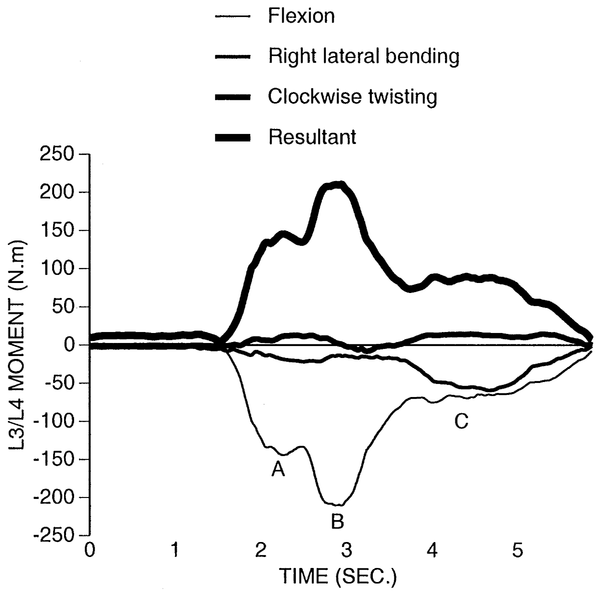

Fig. 2. Example L3/L4 dynamic moment during a lift showing the lift primarily involved generation of extension moments, with smaller lateral bending and twisting moments. The movement begins by flexing forward to grasp the box (time A), lifting the box to waist level (time B), turning to the right to set the box on the stand (time C), and returning to an upright posture (time D).

Table 3
Means (and standard deviation) of low back moments in N m ($n = 22$)

Measure	Average
Peak extension	249 (89)
Mean extension	129 (52)
Peak left lateral bending	45 (20)
Mean left lateral bending	17 (14)
Peak counter clockwise twisting	16 (10)
Peak clockwise twisting	16 (16)
Mean twisting	1 (10)

muscles, as well as for the left external oblique. MIC model predictions were greater for the oblique and latissimus dorsi muscles; SCI model predictions were greater for the erector spinae muscles. The lift mostly involved extension moment with secondary lateral bending and twisting moments (Fig. 2 and Table 3).

4. Discussion

This study was undertaken to explore whether the two optimization models produce substantially different estimates of spinal compression force when analyzing the common industrial task of mildly asymmetric lifting. It was concluded that no biologically meaningful difference in compression force estimates were found for the lifting task studied, which involved mostly sagittal plane moments followed by lateral bending and twisting moments of smaller magnitude.

A limitation of this analysis was the relative simplicity of the mechanical and anatomical representation of the lumbar region. Unlike other models that have multiple vertebral levels, stability requirements, and many muscle fascicles (Stokes and Gardner-Morse, 1995; Cholewicki and McGill, 1996), the models studied here use a single cutting plane and five bilateral pairs of muscles. The anatomic and mechanical simplicity of these models represent a compromise between overly simplistic anatomic representations and sophisticated models having large input data requirements. In the absence of inexpensive and widely available biomechanical job analysis tools having advanced mechanical and anatomical detail, single cutting plane models provide ergonomists with approximate estimates of spinal loads.

The motivation behind this study was to clarify the behavior of a computational model that is often used for ergonomics job analyses. Hughes et al. (1994), Hughes and Chaffin (1995), and McMulkin (1996) have argued that the MIC does not predict electromyographic data collected in the laboratory as well as an SCI model. Moreover, Hughes (1995) showed that there was a 16% difference in predicted spinal compression when analyzing a variety of job tasks found in work at an aluminum smelter, and there was a difference of 3625 N in one specific task involving substantial axial torsion loadings on the low back. These previous studies may lead some to believe that the MIC formulation is inappropriate. The results reported here demonstrate that the two models do not provide substantially different load estimates for mildly asymmetric lifts, so the computationally simpler MIC model does not seriously compromise the analysis. The differences in results between this study and that reported in Hughes (1995) may be accounted for by the different tasks studied. Hughes (1995) analyzed jobs involving significant torsion; this study investigated a task where extension was the predominant component of the low back moment. Therefore, caution is still suggested in using the MIC model when analyzing tasks that require a predominant axial torsion moment.

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