AN EMG ASSISTED BIOMECHANICAL MODEL OF LUMBAR SPINE WITH PASSIVE COMPONENTS

[1] Yu Shu, [1]Judith M. Burnfield, [2]Gary A. Mirka
[1]Madonna Rehabilitation Hospital, Lincoln, NE,
[2]Department of Industrial and Manufacturing Systems Engineering, Iowa State University, Ames, IA
email: yshu@madonna.org, web: http://www.madonna.org/research_institute

INTRODUCTION

In order to accurately predict the load on the spine during manual lifting exertions, a biomechanical model must realistically represent the complicated interactions between various spinal components such as active muscles, ligaments, inter-vertebral discs, and abdominal pressure. Different methods can be used to describe these components - thus electromyography (EMG) assisted, optimization and finite element models have been developed. However, each of these techniques has its own advantages and disadvantages. For example, EMG-assisted biomechanical models are able to capture complicated co-activation patterns of the agonist and antagonist muscles of the trunk. However, models based exclusively on EMG activity are unable to estimate the load from passive components of the extensor mechanism which play a significant role in the net extensor moment, especially near or at full trunk flexion posture. On the other hand, finite element models can precisely simulate the viscoelastic behavior of the passive components of the spine, but estimations of active muscle forces are usually based on assumptions and may not represent the wide range of possible activation patterns in real lifting exertions. The current study demonstrates a novel method to combine these approaches to create a new EMG assisted biomechanical model that includes passive components of the spine.

METHODS

Eighteen subjects voluntarily participated in this study. They performed repetitive controlled and free dynamic lifting and lowering exertions at different combination of speed and load conditions. In the first experiment, six subjects were secured in an isokinetic dynamometer which restrained the pelvis and lower extremity and they performed isokinetic flexion/extension exertions at the speed of 10°/s, 20°/s and 30°/s with 10%, 20% and 30% of their maximum extension force. During the second experiment, twelve subjects slowly lifted and lowered a load weighted 10%, 20% and 30% of the subjects maximum lifting capacity from a squat (bent knee) and stoop (straight knee) posture. In all of these trials, the subjects were asked to reach their full trunk flexion posture. The EMG activity of the major trunk muscles (erector spinae, latissimus dorsi, rectus abdominis, external obliques and internal obliques) and the kinematics of the trunk movements were recorded.

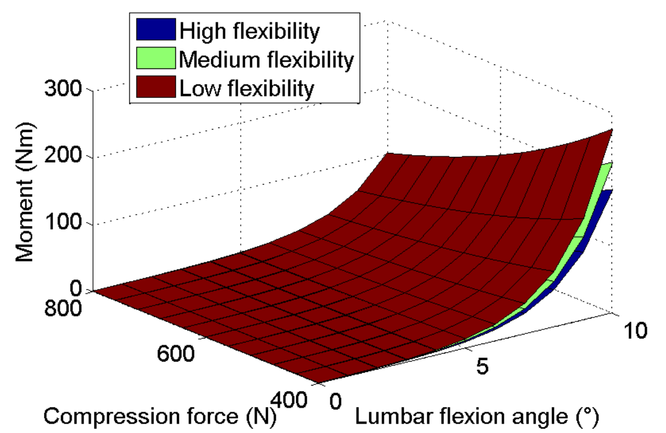

Figure 1: Illustration of the 3D surface of passive moment vs. load and lumbar flexion angle

Passive tissue forces were estimated through the use of a finite element model of the lumbar region [1]. A 3-dimensional response surface of net moment from passive components (Figure 1) was first constructed based on compression force, lumber flexion angle, and subject’s flexibility conditions. The estimated value of the instantaneous passive contribution during the extension/flexion exertion was determined by the instantaneous compression force from active muscles and trunk flexion angle. The forces from active muscles were estimated from normalized EMG activities (multiplied by the estimated cross-sectional area of the muscles and the maximum muscle stress value). The origin and insertion of the muscles were estimated from the anthropometric data of the subjects [2] and were used to calculate the moment arms for active muscles. The total internal moments from both passive components and active muscles were compared with the measured net external moment to validate the model. Two different EMG-assisted models (Model 1: muscle only, without passive components; Model 2: muscle with passive components) were compared with the measured net external moment to provide insight into the utility of the inclusion of these passive tissue forces. Coefficient of correlation (r^2) values between measured and model predicted sagittal moments of the two models in different conditions were calculated to quantitatively assess the difference.

RESULTS AND DISCUSSION

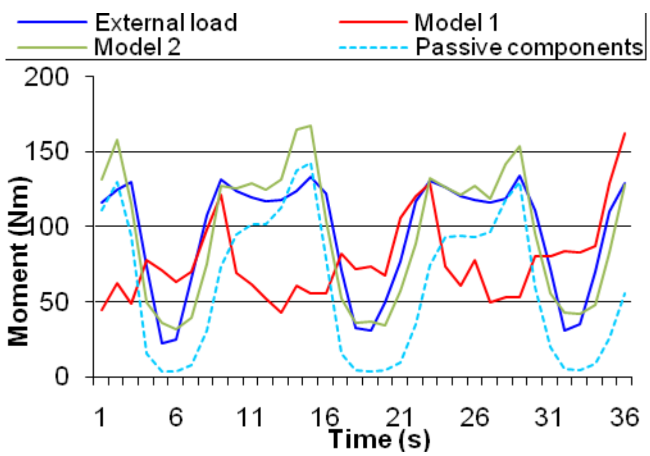

Figure 2: Measured sagittal moment vs. model predicted sagittal moment in three free dynamic lifting/lowering exertions

The sagittal moment relative to L5/S1 joint in three lifting/lowering exertions from one subject is presented in Figure 2. The subject started from a fully flexed posture, lifted the load to the upright posture (about 7 seconds), lowered the load (about 13 seconds), and repeated the lifting motion two more times. The trace from the model 1 (muscle only) showed the flexion relaxation phenomenon with the muscle activity diminishing as the subject reached the full flexion posture. At this posture, passive components generated the majority of the restorative moment to support the trunk. Model 2 (with passive components) correctly predicted the net internal moment by including the moments from passive components.

The necessity of considering the moments generated by passive components in trunk exertions at or near full flexion postures is obvious from the improvement of r^2 values illustrated in Table 1. Additionally, the consistency of these values in different experimental setups indicated the model with passive components was robust across various load, speed, pelvis restraint and knee posture conditions.

CONCLUSIONS

This study introduced a new method for including the effects of passive components of the spine in an EMG-assisted biomechanical model and demonstrated significant improvements in predicting the spinal load at or near full flexion postures.

REFERENCES

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2. Marras WS, et al. *Spine* **20**(13): 1440-1451, 1995.

Table 1: Comparison of r^2 values between measured external sagittal moments and model predicted internal moments for model 1 (muscle only) and model 2 (with passive components)

	Controlled (pelvis restrained)						Free	
	Load			Speed			Knee posture	
	10%	20%	30%	10°/s	20°/s	30°/s	Bend	Straight
Model 1	.47±.15	.34±.25	.28±.20	.38±.22	.40±.22	.32±.20	.12±.13	.12±.11
Model 2	.68±.16	.68±.16	.63±.12	.69±.14	.70±.11	.61±.17	.76±.9	.74±.11