PREDICTION OF TRUNK MUSCLE FORCES AND INTERNAL LOADS DURING FORWARD FLEXION ACTIVITIES

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Knowledge of load distribution among passive and active components of the human trunk during various occupational and sportive activities is essential to assess the risk of injury and to improve prevention, evaluation, and rehabilitation of spinal disorders. To solve the trunk redundancy toward determination of muscle forces and passive loads in forward bending tasks ± loads in hands, a novel synergistic kinematics-based approach coupled with a nonlinear finite element model are introduced. As a part of this study, trunk kinematics needed as input data and surface EMG activity of selected abdominal/back muscles needed for validation of model are measured in normal subjects during isometric forward bending tasks. Predictions are in satisfactory agreement with in vivo measurements. The model proves promising in exercise and rehabilitation applications.

KEY WORDS: spine, muscle, load, posture, finite element.

INTRODUCTION: Injuries to the trunk muscles and ligamentous spine are prevalent in both sportive activities and manual material handling tasks. Lumbar and cervical regions of the spine are particularly at higher risk during various contact and non-contact sports. An accurate estimation of muscle forces and spinal internal loads is of prime importance in sport performance-enhancement programs and in effective prevention, evaluation, and rehabilitation of spinal disorders observed in athletic activities.
Direct in vivo measurement of human muscle and internal spinal forces are almost impossible due to the invasiveness of and difficulty in the procedure. Biomechanical models are recognized as indispensable tools to overcome the existing kinetic redundancy in partitioning net applied moments in order to determine muscle forces and internal passive loads. The existing biomechanical models based on reduction method, EMG-assisted, and optimization satisfy the equilibrium only at one level along the spine. Moreover, the computed muscle forces and internal loads may not at all generate, and hence be compatible with, the posture based on which they were initially calculated. To overcome these major shortcomings, we have recently introduced a synergistic kinematics-based approach in which a-priori measured kinematics of the spine in an activity along with passive and active properties are exploited in a nonlinear finite element model (Kiefer et al., 1998; Shirazi-Adl et al., 2002 and 2004). In the present study this approach is used to compute muscle forces and spinal internal loads during isometric forward flexions of ~35° and ~70° with and without loads held symmetrically in both hands. Moreover, as a part of this investigation in order to obtain input data for the model and validate predictions, trunk kinematics by skin markers and selected extensor/flexor muscle EMG activities by surface electrodes are measured in vivo on normal subjects under same postures and loads.

METHOD: Fifteen healthy male participated in the experimental part of this study. Using a three-camera Optotrack system, a total of 16 skin LED markers were placed to continuously record posture, pelvic tilt, and load location. Five pairs of EMG surface electrodes were placed symmetrically on the left/right sides to record the activity of major superficially located muscle groups; longissimus dorsi, iliocostalis, multifidus, external obliques and rectus abdominis. EMG activity of each muscle was normalized to the maximum EMG observed for that muscle during the MVC tests and subsequent experiments. Under forward bending tasks of ~35° and ~70° with straight knees, loads of 0 N, 90 N and 180N were carried in hands via a bar with arms extended in gravity direction.
A sagittaly-symmetric model of thoracolumbar spine, T1-S1, was used. The nonlinear and direction-dependent mechanical properties of T12-S1 segments were represented by 6 deformable beams while the T1-T12 segments were assumed as a single rigid body. A

sagittaly-symmetric muscle architecture with 46 local muscles (attached to the L1-L5 vertebrae) and 10 global muscles (attached to the thoracic cage T1-T12) were considered (Shirazi-Adl et al., 2004). The trunk gravity load was simulated by a total of 387.1 N distributed eccentrically along the entire length of the spine (Shirazi-Adl et al., 2004). External load of 180 N simulating our in vivo studies with loads carried in hands was also considered in some cases in which it was applied on the T4 level at eccentricity and height measured in experiments.
The novel kinematics-based muscle force evaluation algorithm coupled with an optimization approach (sum of cubed muscle stresses) was subsequently employed to solve for the redundant active-passive system subjected to prescribed measured kinematics and applied gravity and external loads (Shirazi-Adl et al., 2002 and 2004). In an attempt to determine stability margin of the system under calculated muscle forces and gravity/external loads, muscles were replaced by uniaxial elements with different stiffness values assuming proportional force-stiffness relationship (Bergmark, 1989). Nonlinear, linear stability and linear perturbation analyses were carried out in each case to estimate the stability margin.

RESULTS: Local compression and shear forces as well as internal moment at disc mid-heights at different spinal segments computed for standing posture under gravity load alone and two flexed postures (~35° and ~70°) with and without additional external load of 180 N are listed in Table 1. Total muscle force calculated from the kinematics-based algorithm was subsequently decomposed into a passive and an active component using a passive force-change in length relationship (Davis et al., 2003). For the sake of validation, the foregoing active force component was subsequently normalized by the maximum active force at optimal length assuming a maximum stress of 0.6 MPa. Good agreement is found when comparing normalized measured EMG activity of global illiocostalis and longissimus in forward flexion tasks with normalized force predictions, as shown in Fig. 1. Similarly, under similar posture and loading conditions, good agreement was found between the in vivo measured intradiscal pressure values (Wilke et al., 1999) and model predictions using the disc pressure-compression force relationship computed in lumbar motion segment studies (Shirazi-Adl and Drouin, 1988) (Fig. 2). The stability analyses confirmed the stability of the system requiring very small muscle stiffness coefficient values.

Table 1 Internal loads in passive spine at various disc mid-heights under different postures and external loads.

Disc	Standing posture			Forward Flexion 35°						Forward Flexion 70°					
	0 N			0 N			180 N			0 N			180 N		
	*M	*C	*S	M	C	S	M	C	S	M	C	S	M	C	S
T12-L1	8.4	337	-35	18.5	933	250	22.4	1679	490	23.8	1089	407	25.4	1781	672
L1-L2	6.3	405	-46	22.1	1171	224	28.3	2186	439	31.3	1356	375	33.9	2444	637
L2-L3	3.9	447	-63	21.3	1414	111	28.7	2606	202	33.0	1664	225	36.0	3039	338
L3-L4	1.5	498	-7	17.2	1669	173	24.7	2971	277	29.9	1986	257	32.8	3528	365
L4-L5	1.3	535	28	16.8	1862	95	24.1	3250	73	29.2	2245	79	32.0	3859	17
L5-S1	2.3	570	190	19.4	1912	502	27.1	3308	726	31.8	2269	498	35.0	3850	708

* M: sagittal moment, +ve for flexion (N-m); C: local axial compression (N); S: local shear force, +ve in anterior direction (N).

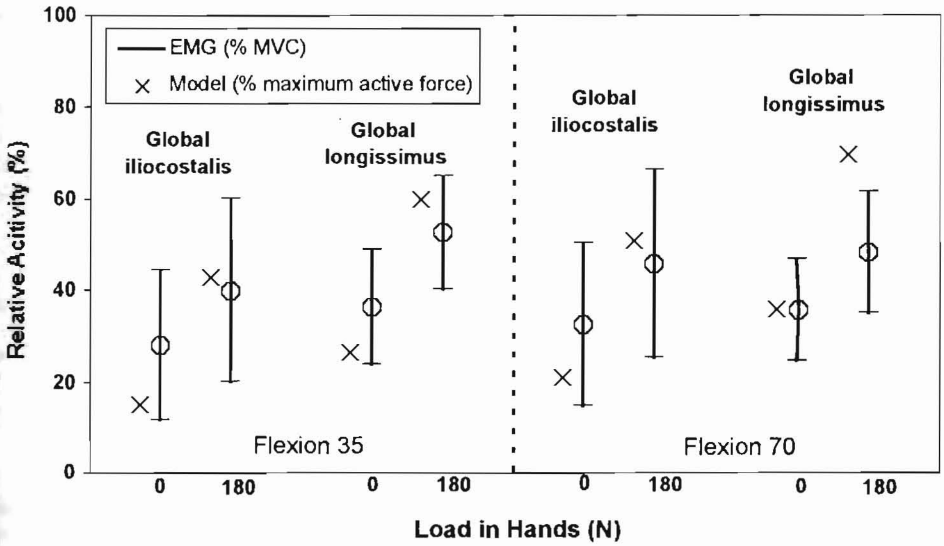

Figure 1: Comparison of measured normalized EMG (mean ± SD) versus computed muscle forces normalized to maximum force (0.6 x PCSA) for global muscles at different postures with and without external load in hands.

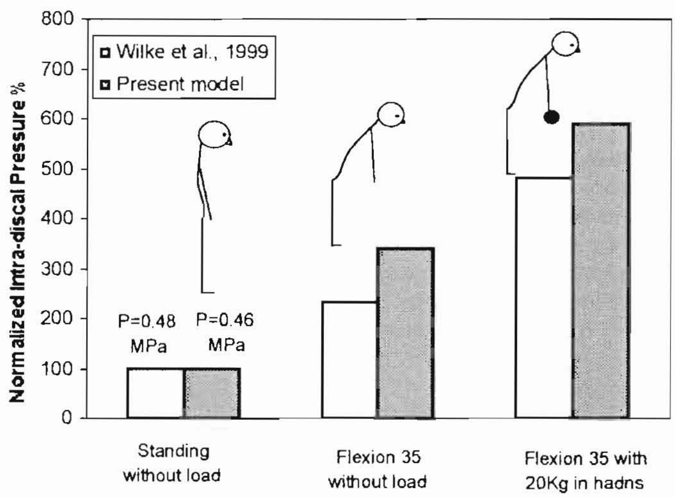

Figure 2: Measured (in vivo) and computed normalized intradiscal pressure values at L4-L5 disc in 35° forward flexion with and without load in hands.

DISCUSSION: The novel kinematics-based approach coupled with nonlinear FE model studies was found as a reliable tool to compute trunk muscle forces and internal loads under forward flexion tasks which is a prevalent posture in many athletic activities and manual material handling tasks. The proposed model accounts for the active-passive synergy by simultaneous consideration of passive ligamentous structure with nonlinear properties and muscle forces at deformed configurations under given postures and loads. The predictions, therefore, satisfied kinematics, equilibrium and stability requirements at all spinal levels and in all directions. As expected, comparing to neutral standing postures with no external load, forward flexion of 35° and 70° and external load of 180 N substantially increased internal loads at all levels and extensor muscle forces (Table 1). Good agreement between and measurements (of this study and those of others) was found for muscle activities and intradiscal pressure values. The spine appears to be rather stable in forward flexion tasks due primarily to the greater stiffness of both active and passive sub-systems that significantly increases with larger flexion angles.

CONCLUSION: The lumbar spine is at high risk of injury in different sports and industrial tasks. The model presented in this study, by synergistic consideration of both active and passive spinal components, appears very promising in accurately predicting trunk muscle forces and spinal internal loads in various activities. This in turn can be used in prevention and treatment of spinal injuries as well as in sport performance-enhancement and rehabilitation programs.

REFERENCES

Bergmark A., (1989) Stability of the lumbar spine -A study in mechanical engineering. Acta Orthopaedica Scandinavia Supplementum. 230, 1-54.

Davis, J., Kaufman, K.R., & Lieber, R.L. (2003). Correlation between active and passive isometric force and intramuscular pressure in the isolated rabbit tibialis anterior muscle. Journal of Biomechanics 36, 505-512.

Kiefer, A., Shirazi-Adl, A., & Parnianpour, M. (1998). Synergy of human spine in neutral postures. European Spine Journal 7: 471-479.

Shirazi-Adl, A., & Drouin, G. (1988). Nonlinear gross response analysis of a lumbar motion segment in combined sagittal loadings. Journal of Biomechanical Engineering 110, 216-222.

Shirazi-Adl, A., Sadouk, S., Parnianpour, M., Pop, D., & El-Rich, M. (2002). Muscle force evaluation and the role of posture in human lumbar spine under compression. European Spine Journal 11, 519-526.

Shirazi-Adl, A., El-Rich, M., Pop, D.G., & Parnianpour, M. (2004). Spinal Muscle forces, internal loads and stability in standing under various postures and loads: applications of kinematics-based algorithm. European Spine Journal, submitted.

Wilke, H.J., Neef, P., Caimi, M., Hoogland, T., & Claes, L.E. (1999). New in vivo measurements of pressures in the intervertebral disc in daily life. Spine, 24, 755-763.

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