COMPARISON BETWEEN EMG-ASSISTED AND EMG-ASSISTED OPTIMIZATION MODELS IN TERMS OF ESTIMATING LUMBAR SPINAL LOADS DURING A GOLF SWING

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This study compared two different models that were required for the computation of the lumbar spinal loads during a golf swing. Simple scaling and optimization techniques were used to modify the initial muscle forces obtained by the EMG-based model such that the condition for moment equilibrium was met. The results indicated that the discrepancies occurred in lumbar spinal loads estimated by the EMG-assisted (EMGA) and EMG-assisted optimization (EMGAO) models due to the differences in gain values obtained by these models. Unlike the EMGA model, the EMGAO model attempted to find optimized gain values for individual muscles. The EMGAO model not only satisfied the moment equilibrium conditions but also determined final muscle forces within a realistic range. Therefore, the EMGAO model would be superior to the EMGA model in terms of muscle force prediction or lumbar spinal loads estimation.

KEY WORDS: musculoskeletal model, lumbar spinal loads, golf swing, EMG-assisted, EMG-assisted optimization

INTRODUCTION: The mathematical or biomechanical model approach has been used for the determination of loads acting on the vertebral column over the past several decades. This approach usually requires the use of many simplifying assumptions about the structure and behavior of the musculoskeletal system in order to make a mathematically determinant model. Thus, most models in the past that were designed to estimate tissue loading have been static, two-dimensional (e.g., Chaffin, 1969, Chaffin & Baker, 1970), and focused only on the sagittal plane (e.g., Schultz, Warwick, Berkson, & Nachemson, 1979). In the past two decades, models that include multiple muscles and ligaments, rather than single equivalents, have been developed (e.g., Anderson, Chaffin, Herrin, & Mathews, 1985; McGill & Norman, 1986). Anatomically detailed biomechanical models, which are designed to estimate a variety of muscle and other tissue forces, must confront the problem of mathematical indeterminacy in which the number of unknowns (forces) exceeds the number of independent force and moment equilibrium equations. To overcome this indeterminacy problem, the surface EMG was commonly used in the partitioning of the total reaction moment acting about a joint into individual musculatures that can produce forces and moments (e.g., EMG-assisted & EMG-assisted optimization models). Although these two previously developed EMG related models have proved to be a useful tool for predicting lumbar spinal load, they have only been applied to lifting with restricted motion and showed similar results. Indeed, these two models have never been tested in a complex motion. If the testing motion is dynamic such as the golf swing, these two models may show different characteristics. Therefore, it was meaningful to compare the two different computational approaches—the EMG-assisted and EMG-assisted optimization models, which were required for the estimation of the lumbar spinal loads during a golf swing.

METHODS: Five male college golfers (age = 19.4±0.9 yrs) served as the subjects.
Data collection: Each subject performed 10 trials (swings) in a laboratory setting and rated his own performance using a 5-point scale at the end of each trial. Four S-VHS camcorders (Panasonic AG455, 60 Hz) were used to record the movement of trunk and lower extremities. Eight reflective markers were placed on the back of the subject in order to estimate the orientations of the middle and lower trunks and musculoskeletal parameters of the trunk musculature during a golf swing. Two AMTI force plates were used to record the ground reaction forces and moments acting on both feet. Ten pairs of surface EMG electrodes with on-site pre-amplification circuitry were placed on skin surfaces to record the activities of the left and right rectus abdominus, external oblique, internal oblique, erector spinae, and latissimus dorsi muscles.

Data reduction: The trial with the highest rating for each subject was selected for analysis. For each trial being analyzed, six critical events were identified from the video recordings: ball address (BA), end of backswing (EB), Middle of downswing (MD), ball impact (BI), middle of follow-through (MF), and end of follow-through (EF). For the purpose of this study, a golf swing was divided into five phases: (1) take away (TA)—from BA to EB, (2) forward swing (FS)—from EB to MD, (3) acceleration (AC)—from MD to BI, (4) early follow-through (EFT)—from BI to MF, and (5) late follow-through (LFT)—from MF to EF.
Lumbar spinal load: The initial muscle force at a given instant was estimated using an EMG-based model (McGill & Norman, 1986). Theoretically, the joint moment (resultant torque) acting on the L5 lumbar vertebra obtained from a free body diagram analysis must be the same as the sum of the moments about L5 due to muscle forces. The initial muscle forces predicted by the EMG-based model were not able to satisfy the condition for moment equilibrium. In this study, simple scaling (McGill & Norman, 1986) and optimization techniques (Cholewicki & McGill, 1994) were used to modify the initial muscle forces obtained by the EMG-based model such that the condition for moment equilibrium was met. Both approaches utilized the physiological recruitment patterns of agonist synergy and antagonist co-contraction obtained from the EMG-based model. The gain terms, which can be obtained using both EMG-assisted (EMGA) and EMG-assisted optimization (EMGAO) approaches, were then used to recalculate the actual 44 muscle forces ($\mathbf{F}_m$). Once $\mathbf{F}_m$ was computed, the lumbar spinal loads ($\mathbf{F}_l$) acting on the L4-L5 motion segment can be estimated as:

$$\mathbf{F}_l = \mathbf{F}_r - \sum_{m=1}^{44} \mathbf{F}_m \qquad (1)$$

where $\mathbf{F}_r$ is the joint resultant force at L5 level.

EMGA model: The purpose of the EMGA model was to satisfy the moment equilibrium conditions by applying a fixed set of gain values to all initial muscle forces. A simple scaling technique was used to determine the gain values:

$$g_x \sum_{i=1}^{44} M_{xi} = \mathrm{M}_x \qquad g_y \sum_{i=1}^{44} M_{yi} = \mathrm{M}_y \qquad g_z \sum_{i=1}^{44} M_{zi} = \mathrm{M}_z \qquad (2)$$

$$0 < g_x < 10,\ 0 < g_y < 10, \text{ and } 0 < g_z < 10, \qquad (3)$$

where g_x, g_y, and g_z are the gains, M_{xi}, M_{yi}, and M_{zi} are the components of the resultant moment about the X, Y, and Z axes, respectively, due to the ith muscle force, and M_x, M_y, and M_z are components of the joint resultant moment about L5.
The computation (Equations 2-3) was repeated every 1/60 second to find a proper gain for each component at a give time constant. Therefore, a fixed set of gain values was applied to all muscles. Equation 3 was a constraint which restricted the gain value to a fixed range. If the computed gain was out of the range at a certain instant the initial muscle force obtained from the EMG-based model was used as the final muscle force.

EMGAO model: The goal of the EMGAO was to satisfy the condition for moment equilibrium by applying the least possible adjustment to individual muscle forces. In this study, the cost function of the EMGAO was formulated as:

$$\text{Minimizing } \sum_{i=1}^{44} M_i (1-g_i)^2 \quad \text{where } M_i = \sqrt{M_{xi}^2 + M_{yi}^2 + M_{zi}^2} \qquad (4)$$

and subjected to the following constraints:

$$\sum_{i=1}^{44} g_i M_{xi} = \mathrm{M}_x \qquad \sum_{i=1}^{44} g_i M_{yi} = \mathrm{M}_y \qquad \sum_{i=1}^{44} g_i M_{zi} = \mathrm{M}_z \qquad (5)$$

$$g_i \geq 0,\ i = 1,2,\ldots,44 \qquad (6)$$

The optimization (Equations 4-6) was repeated every 1/60 second to compute optimum gain values for individual muscles at a given time instant. The squared term in the cost function (Equation 4) assured that the gains were not concentrated on one or just a few muscles. This penalty term forced the gain to spread evenly among muscles. The last constraint (Equation 6) ensured that all muscle forces were positive.

Data analysis. For each phase of a golf swing, mean and SD values were determined for the maximum and minimum antero-posterior (A/P) shear and medio-lateral (M/L) shear lumbar spinal loads (expressed as % body weight), and average and maximum compressive loads estimated by the EMGA and EMGAO models. One-way ANOVA with repeated measures were performed to test for significant differences ($p < .05$) in the average, maximum, and minimum resultant lumbar spinal loads (three components) between the EMGA and EMGAO model approaches in different phases.

RESULTS AND DISCUSSION: The mean and SD values of the average, maximum, and minimum compressive and shear forces for different phases of a golf swing estimated by the EMGA and EMGAO models are listed in Table 1.

Table 1 Means and Standard Deviations of Average and Peak Compressive Lumbar Spinal Loads for Different Phases of a Golf Swing

	Compressive load				Antero-posterior shear load				Medio-lateral shear load			
	EMGA		EMGAO		EMGA		EMGAO		EMGA		EMGAO	
Phase	Ave	Max	Ave	Max	Max	Min	Max	Min	Max	Min	Max	Min
TA	-255.1*	-831.0	-368.9*	-476.5	130.8**	23.4**	40.8**	-20.9**	332.9	-49.8*	57.1	-4.8*
	(87.3)	(640.6)	(46.9)	(71.8)	(47.1)	(2.9)	(11.8)	(6.9)	(435.7)	(39.9)	(27.1)	(8.8)
FS	-1018.2**	-2786.0*	-411.4**	-653.3*	233.3	30.8	80.3	1.1	126.4	-142.5	44.3	-23.1
	(247.2)	(1357.2)	(44.5)	(100.0)	(186.5)	(9.2)	(24.9)	(29.3)	(141.8)	(197.2)	(23.5)	(24.5)
ACC	-1339.1	-2054.0	-604.7	-629.4	254.5	100.3*	69.6	54.7*	-36.5	-157.9	-3.5	-39.9
	(1283.7)	(1609.9)	(126.1)	(145.7)	(184.6)	(22.3)	(37.9)	(34.9)	(186.3)	(401.2)	(41.7)	(42.8)
EFT	-932.6	-1402.0	-531.1	-557.1	110.3	88.0	96.7	70.7	-13.3	-40.8	-43.7	-60.2
	(368.7)	(792.0)	(81.0)	(93.5)	(24.2)	(13.1)	(34.8)	(28.0)	(61.0)	(82.6)	(55.9)	(63.3)
LFT	-472.5	-1660.6	-297.9	-506.5	337.6	21.4*	163.6	45.2*	42.6	-94.4	30.4	-49.4
	(263.8)	(1059.9)	(46.0)	(88.6)	(195.8)	(14.8)	(35.3)	(12.4)	(53.0)	(84.8)	(13.5)	(64.0)

Note. Unit in % body weight (BW). Significant difference between the two models at *$p < .05$ and **$p < .01$. Standard deviations in parentheses.

Compressive load: Significant differences between the EMGA and EMGAO models were found in the average compressive loads during the take away ($F(1, 4) = 9.82$, $p < .05$) and forward swing ($F(1, 4) = 30.59$, $p < .01$) phases. A significant difference was also found in the maximum compressive load for the forward swing phase ($F(1, 4) = 12.15$, $p < .05$). Unlike the A/P shear load in the take away phase, the average compressive loads estimated by the EMGAO model were significantly larger than the corresponding values from the EMGA model. During the take away phase, the mean average compression estimated by the EMGA model was 40% smaller than that of the EMGAO model. The discrepancies were due to the differences in gain values used in these two models. The mean compressive loads during the forward swing phase predicted by the EMGA model were significantly larger than that of the EMGAO model. This is most likely due to the fixed gain values used in the EMGA model regardless of the differences in the physiological cross sectional area (PCSA) or initial muscle force. Consequently, a muscle with large PCSA or high initial muscle force would have greater force adjustment than a muscle of small PCSA or low initial muscle force. Despite the huge discrepancies between the two models in the load values, both models predicted the maximum compressions in the forward swing phase.

A/P shear load: During the take away phase, significant differences between the EMGA and EMGAO models were found in the maximum A/P shear ($F(1, 4) = 22.31$, $p < .01$) and minimum A/P shear ($F(1, 4) = 248.94$, $p < .01$) loads. Significant differences between the two models were also found in the minimum A/P shear loads in the acceleration phase ($F(1, 4) = 9.52$, $p < .05$) and late follow-through phase ($F(1, 4) = 18.36$, $p < .05$). In the take away phase, the mean A/P shear load estimated using the EMGA model was approximately three times the corresponding values for the EMGAO model. The mean minimum A/P shear load in the acceleration phase obtained using the EMGA model was about twice that of the EMGAO model.

The anterior shear loads obtained from the EMGA model were always greater than the corresponding values from the EMGAO model during the backswing. The discrepancies were likely due to the improper gain values for the EMGA model. At certain events, the gains were negative or beyond the range of constraint so the model was forced to use the initial muscle forces. In addition, because the trunk was mainly twisted in the CW direction (in the overhead view) with lateral bending to the left during the take away phase, the muscles can generate large passive tension in the A/P direction. Therefore, it was possible that the large passive tension predicted by the EMGA model with fixed gain might increase the A/P shear load drastically. This passive tension was particularly noticeable near EB—at the extreme range of motion. The maximum A/P shear loads predicted by the two models in the late follow-through phase indicated that the hyperextension of the trunk (arching back) had a major role in increasing the passive tensions.

M/L shear load: Significant differences between the EMGA and EMGAO models were found in the minimum (left) M/L shear load ($F(1, 4) = 8.97$, $p < .05$) in the take away phase. The EMGA model estimated significantly larger minimum M/L shear loads than the EMGAO model in this phase. Both models predicted maximum (right) M/L shear loads in the take away phase which were 333% BW and 57% BW for the EMGA and EMGAO models, respectively. For the overall minimum M/L shear load, -158% BW was found using the EMGA model in the acceleration phase and -60% BW was observed in the early follow-through phase using the EMGAO model. During the take away phase, the trunk laterally bent to the left with the right axial rotation (lumbar coupling). This left lateral bending generated a large positive M/L shear load (shear load to the right) in this period. With the opposite lumbar coupling, the left axial rotation (CCW direction in an overhead view) results in the right lateral bending, a large negative M/L shear load (shear load to the left) was observed in the acceleration and early follow-through phases.

CONCLUSION: The present study indicated that the discrepancies occurred in lumbar spinal loads estimated by the EMGA and EMGAO models due to the differences in gain values obtained by these models. The EMGA model used three gain values, one for each moment component to determine the final muscle forces. In other words, the three gains were applied to each muscle such that the moment equilibrium conditions were satisfied. Because the same gains were applied to all muscles, a muscle with large PCSA or high initial muscle force would have greater force adjustment than a muscle of small PCSA or low initial muscle force. When the gains are greater than one, as in the case of the present study, unrealistically large muscle forces were often predicted. However, unlike the EMGA model, the EMGAO model attempted to find optimized gain values for individual muscles. The squared term in the cost function and the penalty term forced the adjustments to be spread evenly among different muscles. Using individualized gain values, the EMGAO model not only satisfied the moment equilibrium conditions but also determined final muscle forces within a realistic range. Therefore, the EMGAO model would be superior to the EMGA model in terms of muscle force prediction or lumbar spinal loads estimation.

REFERENCES:

Anderson, C. K., Chaffin, D. B., Herrin, C. D., & Mathews, L. S. (1985). A biomechanical model of the lumbasacral joint during lifting activities. *Journal of Biomechanics*, **18**, 571-584.

Chaffin, D. B. (1969). Computerized biomechanical models-Development of and use in studying gross body actions. *Journal of Biomechanics*, **2**, 429-441.

Chaffin, D. & Baker, W. (1970). A biomechanical model for analysis of symmetric sagittal plane lifting. *AIIE Transaction*, **11**, 16-25.

Cholewicki, J., & McGill, S. M. (1994). EMG assisted optimization: a hybrid approach for estimating muscle forces in an indeterminate biomechanical model. *Journal of Biomechanics,* **27**, 1287-1289.

McGill, S. M., & Norman, R. W. (1986). Partitioning of the L4/L5 dynamic moment into disc, ligamentous, and muscular component during lifting. *Spine*, **11**, 666-678.

Schultz, A. B., Warwick, D. N., Berkson, M. H., & Nachemson, A. (1979). Mechanical properties of the human lumbar spine motion segments: Part I. Responses to flexion, extension, lateral bending and torsion. *ASME Journal of Biomechanical Engineering*, **101**, 46-52.