ESTIMATING LUMBAR SPINAL LOADS DURING A GOLF SWING USING AN EMG-ASSISTED OPTIMIZATION MODEL APPROACH

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This study estimated the lumbar spinal loads at L4-L5 level during a golf swing using musculoskeletal modeling techniques. Data were collected from five college golfers. Four S-VHS camcorders and two force plates were used to obtain 3-D kinetics and kinematics of the golfer's motion and 10 surface electrodes were used to record the activity of selected lower trunk muscles. The EMG-assisted optimization model was used to estimate lower trunk muscle forces. The results demonstrate that the L4-L5 motion segment is subjected to considerable compressive (4,300 N), antero-posterior (A/P) shear (882 N), and medio-lateral (M/L) shear loads (-252 N) during a golf swing. The repetitive changing direction of the shear load during a golf swing may increase the chance of fatigue fracture of pars interarticularis. The results also suggest that a golf swing with shortened backswing and follow-through may reduce the risk of back injury.

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

INTRODUCTION: A number of surveys have shown that the golf swing is increasingly being recognized as a potential cause of lower back injuries. Although many researchers have been interested in seeking a "perfect" golf swing to improve performance, surprisingly, few studies have investigated the potential causes of injury such as back pain. The forces generated by the musculature surrounding the lumbar spine often contribute a large part of spinal stresses. Tissue damage occurs when the force to which it is exposed to at a particular instant exceeds the tissue's tolerance. Therefore, large mechanical stresses on the lumbar spine are widely believed to be related to the development of low-back pain. During a golf swing, the L4-L5 disc as well as the lumbosacral (L5-S1) disc are subjected to large magnitudes of compressive loads in addition to the continuous static load due to the weight of the trunk. In order to explore the relationship between lumbar spinal loads and lower back injury, it is important to have knowledge of the lumbar spinal loads during a golf swing.
The purpose of this study was to estimate the loads acting on the spinal motion segment at the L4-L5 level (lower back) during a golf swing using an electromyography (EMG)-assisted optimization model (Cholewicki & McGill, 1994). It was expected that a golf swing would introduce large compressive, antero-posterior (A/P) shear, and medio-lateral (M/L) shear loads acting on the L4-L5 disc.

METHODS: Five male college golfers (age = 19.4±0.9 yrs, handicap = 0.8 ± 1.1) served as the subjects. Each subject used his own driver (1-wood) and wore his own golf shoes during the tests.
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 (Liberty Technology MYO115 and Therapeutic Unlimited D100) were placed on skin surfaces to record the activity of the left and right rectus abdominus, external oblique, internal oblique, erector spinae, and latissimus dorsi muscles.

The upper and lower four markers were used to define the middle and lower trunk reference frames, respectively. A calibration frame (PEAK Performance Technologies, USA, 25 control points, 2.2 x 1.9 x 1.6 m^3) was videotaped prior to the trials for 3-D space reconstruction purposes. An event synchronization unit (PEAK Performance Technologies) and a microphone were used to synchronize the video, EMG, and force plate recordings. At the instant of impact during a golf swing, the sound captured by the microphone activated the event synchronization unit. As a result, two light emitting diodes (LEDs), which were visible to all camcorders, were turned on and a 3 V signal was forwarded to the A/D converters of both EMG and force plates data collection systems simultaneously.

Data reduction. The trial with the highest rating for each subject was selected for analysis. For each trial being analyzed, six critical instants were identified from the video recordings: (1) ball address (BA)—initiation of backswing, (2) end of backswing (EB)—beginning of the downswing, (3) middle of downswing (MD)—the club at the horizontal position during the downswing, (4) ball impact (BI)—the instant of ball/driver impact, (5) middle of follow-through (MF)—the club at the horizontal position after impact, and (6) end of follow-through (EF)—the instant the club stopped its motion momentarily. For the purpose of this study, a golf swing was divided into five phases: (1) take away—from BA to EB, (2) forward swing—from EB to MD, (3) acceleration—from MD to BI, (4) early follow-through—from BI to MF, and (5) late follow-through—from MF to EF.

Two-dimensional coordinates of 10 body landmarks and eight reference markers were extracted from video images using a Peak Motion Measurement System for each selected trial. The 10 digitized body landmarks were the right and left hips, knees, ankles, heels, and toes. Using the KWON3D motion analysis software (V-tech, Korea), the Direct Linear Transformation (DLT) technique (Abdel-Aziz & Karara, 1971) was used to obtain 3-D coordinates of landmarks and markers relative to the reference frame defined by the calibration frame. The 3-D coordinates were smoothed and transformed to a global reference frame with principle axes parallel to the A/P, M/L, and vertical directions. Considering a free body diagram of the human body below the L5 level, the resultant force ($\mathbf{F}_r$) acting on the L5 level was computed using the known segmental kinematic and inertial characteristics, and the ground reaction forces and moments.

The musculoskeletal parameters used in this study were based on the models of McGill (1992) and Cholewicki (1993). The musculoskeletal parameters included physiological cross-sectional area (PCSA) values of 22 muscles, 3-D coordinates of the origins and insertions of 22 muscles, and the proximal ends of individual lumbar vertebral bodies. The 22 muscles were the right rectus abdominus, external obliques (two parts), internal obliques (two parts), transverse abdominus, pars lumborum (five parts), quadratus lumborum, mulifidus (two parts), illiocostalis lumborum, longissimus thorasis, latissimus dorsi, and psoas (five parts). The musculoskeletal parameters of the corresponding muscles on the left side were obtained as the mirror image of the right side.

The muscle force at a given instant ($\mathbf{F}_m$) was determined using a modification of a model from McGill and Norman (1986):

$$\mathbf{F}_m = g_m[(\sigma_m \mathbf{NEMG}_m A_m \Omega \delta + F_{pec})]\mathbf{e}_m$$

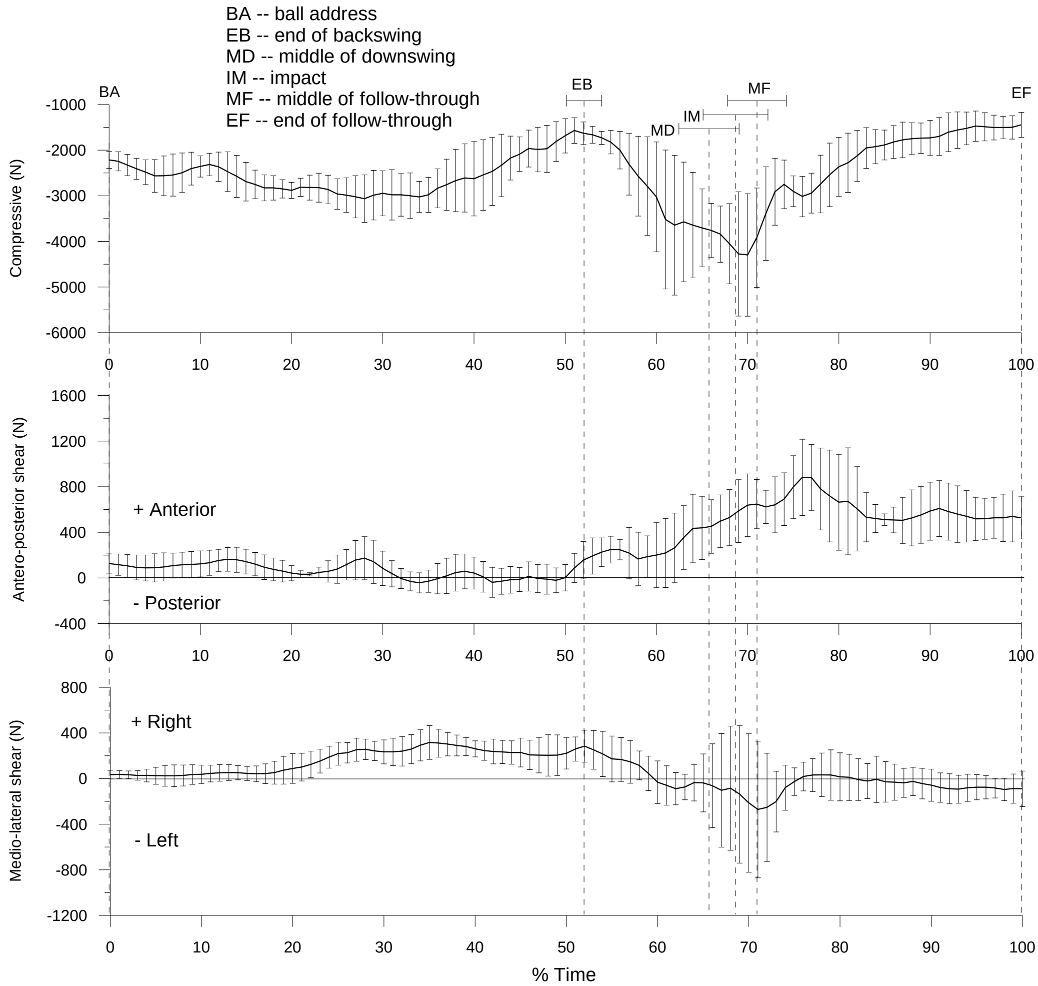
where g_m is a gain term, σ_m is the muscle stress which was set at 35 $N \cdot cm^{-2}$, $\mathbf{NEMG}_m$ is the normalized EMG data, A_m is the PCSA of the muscle (cm^2), Ω is the coefficient of velocity modulation, δ is the coefficient of active length modulation, F_{pec} is the force due to passive elasticity, and $\mathbf{e}_m$ is a unit vector representing the line of action of the muscle. The gain term was obtained using an EMG-assisted optimization approach (Cholewicki & McGill, 1994). The contact force ($\mathbf{F}_c$) acting on the L4-L5 motion segment was estimated using the known $\mathbf{F}_m$ and $\mathbf{F}_r$ values:

$$\mathbf{F}_r = \mathbf{F}_c + \sum_{m=1}^{44} \mathbf{F}_m$$

Knowing the orientation of the L5 vertebra, $\mathbf{F}_r$ was resolved into compressive, A/P shear, and

M/L shear forces.

Data analysis. The lumbar spinal loads were normalized to the duration of a golf swing and mean and standard deviation values for all subjects were computed. In addition to absolute force values, loads were also normalized to the body weight (BW) of the subject.



RESULTS AND DISCUSSION: The lumbar spinal loads at the L4-L5 level during a golf swing are shown in Figure 1. The duration of a golf swing (from BA to EF) was normalized to 100% time. The dashed vertical lines represent the average times for different critical instants and the standard deviations are indicated as horizontal bars at the top of these lines.

Figure 1 - Mean compressive and shear loads acting on the L4-L5 during a golf swing. The error bars represent standard deviations.

Compressive load. The mean average axial compression during the take away phase was about 370% BW. The compressive load increased steadily after EB and reached its maximum of 605% BW (or an average of approximately 4,300 N) near IM. The mean average compressive loads were 531% BW and 298% BW for the early and late follow-through phases, respectively. The minimum mean compression of 202% BW (about 1,440 N) was recorded at EF. The peak compressive load estimated in the present study was about 700% BW. However, Hosea et al. (1990) reported that both professional and amateur golfers generated peak compressive loads greater than 800% BW. They also found two major peaks and one minor peak in the compressive loading pattern during a golf swing. This three-peak pattern was not observed in the present study.

A/P shear load. The A/P shear loads were relatively small (≅ 64 N) during the backswing but the mean A/P shear load increased gradually after EB and reached its peak of 124% BW (882

N) at around 75% swing time (duration). The present study showed relatively large anterior shear loads throughout a golf swing except the small posterior shear loads (≅ -21 N) between 30 to 50% duration. An anterior shear load tends to displace the L4-L5 motion segment forward relative to the middle trunk. On the contrary, Hosea et al. (1990) reported large posterior shear loads during the forward swing and acceleration phases and the peak posterior shear load (≅ -596 N) was recorded near IM. The differences could be due to the difference in modeling techniques. However, the hyperextension of the trunk, which was clearly observed during the follow-through phases, could produce large anterior shear loads. Thus, the results from the present study are more reasonable and accurate than those reported by Hosea et al.

M/L shear load. The mean M/L shear load was mostly acting toward the right during the take away phase. The direction was reversed during the early downswing and reached its peak (-252 N) around MF. Relatively small M/L shear loads were observed during the late follow-through phase. Hosea et al. (1990) reported that professional golfers generated a mean peak M/L shear load (shear load to the right) of 530 N while amateur golfers produced a mean peak of 960 N. In their study, the amateurs exhibited a peak right M/L shear load during the forward swing phase and a peak left shear load in the acceleration phase. The two-peak loading pattern was also found in the professionals—a peak right M/L shear load near IM and a peak left shear load during the early follow-through phase. The present study predicted a mean peak right M/L shear load of 404 N during the take away phase and a mean peak left M/L shear load of 463 N during the early follow-through phase. However, the times of occurrence for the peak M/L shear loads were not comparable to those values reported by Hosea et al.

CONCLUSION: The present study demonstrated that the L4-L5 motion segment is subjected to considerable compressive, A/P shear, and M/L shear loads during a golf swing. However, the loads from a single swing seldom fall within the range of damaging loads. Thus, it seems that the magnitude of the load is not the primary factor for causing lower back injury. The repetitive changing direction of the shear load during a golf swing may increase the chance of fatigue fracture of pars interarticularis. In addition, the lumbar vertebra and disc are made of biological materials and the disc is viscoelastic in nature and consequently time rate dependent. Therefore, accumulated stress due to repeated golf swings may lead to disc degeneration, and even submaximal exertions may lead to structural deformation of the lumbar spine. A golf swing with shortened backswing and follow-through may reduce the risk of back injury because the smaller range of motion of the trunk reduces passive tensions of the lower trunk muscles and lumbar spinal loads.

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