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Evaluation of Spinal Internal Loads and Lumbar Curvature under Holding Static Load at Different Trunk and Knee Positions

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Abstract: A study was performed to investigate how different trunk and knee positions while holding static loads affect the lumbar curvature and internal loads on the lumbar spine at L_4-L_5. Ten healthy male subjects participated in this study. Two inclinometers were used to evaluate the curvature of lumbar spine, lordosis, while a 3D static biomechanical model was used to predict the spinal compression and shear forces at L_4-L_5. Eighteen static tasks while holding three level of load (0, 10 and 20 kg), two levels of knee position (45 and 180° of flexion) and three levels of trunk position (neutral, 15 and 30 degree of flexion) were simulated for 10 healthy male subjects. The results of this study revealed that the lordosis of lumbar spine changed to kyphosis with increasing weight of load from 0 to 20 kg in trunk flexion position ($p<0.05$), but in squatting position (45° knee full flexion) the higher load did not affect the curvature. The results of this study suggested, at a more flexed trunk and standing position with higher loads both external moment and internal loads increased significantly at L_4-L_5 level but with 45 knee flexion external moment and compression force increased and shear force decreased significantly ($p<0.05$). Subjects made more effort to maintain stability of the body in squat position. The highest external moment and compression force were computed at flexed knee and trunk position with highest loads. Hence holding weight in this position must be avoided by implementing ergonomic change to the workplace.

Key words: Posture, trunk and knee position, holding static tasks, internal loading

INTRODUCTION

Despite the high prevalence of the Low Back Pain (LBP), knowledge of its cause, methods of its prevention and treatment remain unclear (Bogduk and Twomy, 1992; Chen Wen-Jer *et al*., 1998; Parnianpour *et al*., 1997). Epidemiologic studies have associated heavy physical work, static work postures, frequent bending and lifting with LBP. Review of literature have shown lifting accounted for 33% of all work-related causes of back pain (Arjmand and Shirazi-Adi, 2005).

Minimal changes in lumbar posture (posterior pelvic tilt and lumbar flattening) substantially influenced muscle force, internal loads and margin of spine stability (McGill *et al*., 2000; Shirazi-Adl *et al*., 2005). It has been suggested the relationship between the lordosis of the lumbar spine and trunk and knee angles may be constrained by the necessity to optimize stress distribution in the osteoligamentous lumbar spine. (Mitnitski *et al*., 1998; Anderson *et al*., 1986; Lee and Chen, 2000; David and Whittle *et al*., 1996; Shirai-Adl *et al*., 2002).

Despite the well recognized role of lifting in low back injuries, the literature on safe lifting techniques remains controversial. A kyphotic lift is recommended by some as it uses the passive ligament system and then relieving the active extensor muscles. In contrast others advocate straight- back and flexed knee posture indicating that posterior ligaments cannot effectively protect the spine against internal loads (compression and shear forces) on the spine during sudden loading or in presence of perturbations (Burgress-Limerick, 2003; Straker, 2003; Shirazi-Adl *et al*., 2005).

Some the past literature have evaluated the effects of biomechanical parameters on lifting from point of view vertebral orientation or lumbar posture, without considering internal loads (Lee, 2004; Akira and Tatsuro *et al*., 1995; Eklund and Liew, 1991). While some others have studied electromyography (EMG) of trunk or hip muscles the lumbar posture during holding or lifting tasks. (Shin *et al*., 2004; Vakos *et al*., 1994). Recently, novel techniques have used EMG-assisted models evaluating lumbar posture and internal load on spine in lifting tasks (Arjmand and Shirazi-Adl, 2005, 2006; Shirazi-

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Adl *et al*., 2004; El-Rich *et al*., 2004). However use of EMG -assisted models in industrial setting is rather limited since these models require extensive expertise and complex experimental setups not available to most ergonomics. Due to these limitations, biomechanical models (based optimization techniques) have been recognized as indispensable tools for evaluation of spinal loads and system stability in lifting tasks (Chaffin and Anderson, 1999).

The purpose of the current study was to determine the effects of the external load, trunk and knee flexion position on lumbar curvature and internal spinal loading which are crucial in understanding the risk of back injuries in holding static tasks.

MATERIALS AND METHODS

Subjects: Ten healthy male subjects volunteered to participate in this experiment. Mean±SD age was 23.2±1.3 years, mean structure and weight were 176.6±4.27 cm and 62.5±5.2 kg, respectively. A questionnaire was administered to each subject to ensure that there was no history of serious back or knee disorders and other musculoskeletal disorders.

The experiment consisted of three-way within-subject designs. The independent variables included: trunk posture, knee posture and weight handled. The sagittaly symmetric lifts were chosen to replicate closely the kinematics conditions observed in the low- risk groups of a companion industrial study (Fathallah and Marras, 1998).

Trunk posture was set at three levels: Neutral, 15° and 30° flexion from gravity line. The neutral posture was selected instead of 0° flexion angle, because some tasks absolutely could not accomplish in 0° trunk flexion and also the neutral posture usually were used more in industrial situations. The knee posture was chosen in two levels to replicate the posture patterns recommended to healthy and occupational LBP patients during rehabilitation: 180° flexion (standing posture) and 45° flexion (squat posture). Three weight level was considered low (0 kg), medium (10 kg) and high (20 kg). The dependent variables consisted of the lumbar curvature and computed sagittal external moment, compression and anterior-posterior shear forces at L_4-L_5.

Instruments: Lumbar curvature and also static trunk posture were recorded using two small inclinometers attached to the skin overlying the spinous processes at T_{11} and S_1 (Model 0725, Fredericks Co., USA) (Akira and Tatsuro, 1995; Dolan *et al*., 1988). The full descriptions of the developed biofeedback technique and sensor calibration were detailed elsewhere (Kahrizi *et al*., 2004). The difference between two sensor's position were computed as spine curvature. The values of deflection from the vertical line toward flexion were assigned positive values (kyphosis) whereas those toward extension were assigned negative ones (lordosis).

During the experiment, the subjects held the load of 10 or 20 kg in a box, for load 0 kg., subjects held the paper box of 0.370 kg with similar shape and size to first box. The dimension of the both boxes were 26×30×40 cm with two handles (4.3 cm in diameter and 11.5 cm in length) centered at their sides. During all tasks the moment arm of external load were constant for each subject on basis of his anthropometric characteristics via using a stationary, moment arm frame that were designed for this study (Kahrizi *et al*., 2004).

Experimental protocol: Each subject was asked to maintain the posture in one of the 18 tasks randomly assigned holding the loaded box (Fig. 1) while maintaining the moment arm using the moment arm frame mentioned earlier. Each subject repeated each tasks for three times with one minute rest in between to avoid any fatigue.

A mathematical model developed by University of Michigan, 3 Dimensional Static Strength Prediction Program (3DSSPP), was used to predict the external moment and net compression and shear forces by the ten muscles included at the level of L4-L5. This program enables a designer to simulate a large variety of work

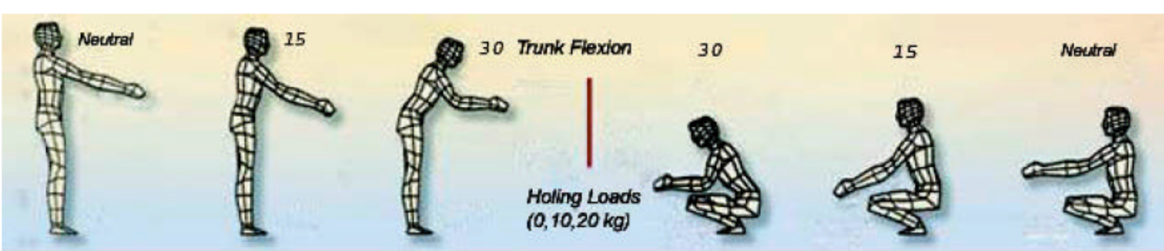

Fig. 1: The eighteen holding tasks consisted of holding three levels of load in these six different trunk and knee positions

tasks and evaluate the effects of gender, posture and external load on spinal loads, required muscular strength and percentage of population capable of carrying the static 3D exertions (Chaffin and Anderson, 1999).

Statistical analysis: Dependent variables of external moment, internal loads (compression and anterior-posterior) and averaged lumbar curvature were examined using a repeated- measures ANOVA to test the main and interaction effects of load, trunk and knee postions. Also, Post- hoc analysis were performed to examine differences between levels of significant independent variables at significant of $\alpha = 0.05$.

RESULTS

The descriptive results of the measured and computed dependent variables are presented for each of the 18 experimental conditions in Table 1 while the p-values of ANOVA testing the main and interaction effects of load, trunk and knee positions are presented in Table 2. The computed shear force was the most sensitive measure to the experimental conditions (Table 2). The predicted shear force increased at higher load, more trunk flexion and in fully extended knee position (i.e., standing posture). The effect trunk flexion was much stronger on shear force in standing posture than in squatting position

Table 1: Predicted External Moment, Internal loads at L_4-L_5 level and Lumbar Curvature for the 18 Experimental Static Holding Tasks

	Knee 180°								
	Load (kg)								
	Trunk °								
	N			15°			30°		
	0	10	20	0	10	20	0	10	20
External moment (Nm)	-14±4.8	-17.0±9.59	-29.76±7.76	29.06±604	-36.25±7.14	-43.93±6.19	-53.39±7.26	58.34±6.33	-64.13±7
Compression Force (N)	336.2±59.8	560.7±139.1	765.4±125.6	744.3±119.5	855.8±131.8	988±114.5	1108.7±131	1195.8±118.6	1295.5±127.7
A-P Shear Force (N)	74.6±20.75	92.7±78	118.8±10.62	121.6±11.7	125±11.42	129.3±10.5	148.6±16.24	153±15.2	158.8±16.4
Lumbar Curvature (°)	-13.73±3.9	-7.63±6.82	-6.11±7.6	-1.6±4.4	-0.88±5.52	2.09±4.9	7.44±7.6	8.84±9.6	12.25±9
	Knee 45°								
	Load (kg)								
	Trunk °								
	N			15°			30°		
	0	10	20	0	10	20	0	10	20
External moment (Nm)	-34.85±16	-38.23±15.8	50.91±12.8	-38.49±7.14	-44.61±9.31	-51.31±8.27	-65.25±9.67	-69.29±10.92	-77.46±13.64
Compression Force (N)	816.9±288.6	934.40±278.1	1154.7±234.4	927±149.4	1041.5±174.3	1165±154.7	1386±184.5	1469.7±206	1614±252.4
A-P Shear Force (N)	63.1±9.89	65.2±11.4	75±10.26	68.3±7.3	71±7.8	73.6±6.6	82.4±10.2	84.4±11.7	90.5±14.25
Lumbar Curvature (°)	4.0±10.5	2.49±10.48	5.74±11.9	6.27±7	7.7±5.5	8.12±8.35	21.48±12.65	19.7±11	21.14±11

Table 2: p-values of the ANOVA results of main and interaction effects of load, trunk and knee position on lumbar curvature and predicted external moment and internal loads (T: = Trunk position, L: = Load, K: = Knee position)

p-value of ANOVA	Dependent variables: External moment (Nm)	Compression force (N)	Shear force (N)	Lumbar curvature (°)
Trunk position (Neural, 15°, 30°)	0.000	0.000	0.000	0.000
Load (0, 10 and 20 kg)	0.000	0.000	0.000	0.091
Knee position (180°, 45°)	0.000	0.000	0.000	0.000
T×L	0.145	0.235	0.002	0.915
T×K	0.000	0.002	0.000	0.325
L×K	0.762	0.857	0.002	0.207
T×L×K	0.886	0.915	0.032	0.772

(T×K effect was significant, ($p<0.000$). As expected the predicted response of external moment and compression force were similar. The compressive force and external moment were higher at higher load, more flexed trunk position and in the squatting position. Lumbar curvature was only significantly affected by the main effects of trunk and knee positions while the main effect of load was approaching significant level ($p = 0.09$). Lumbar curvature became more kyphotic at the higher loads irrespective of what the lumbar or knee positions were (the interaction effects of L×K and T×L on lumbar curvature were insignificant, Table 2).

MEASUREMENT OF LUMBAR CURVATURE

The results of the ANOVA for the magnitudes of lumbar curvature revealed that main effects knee and torso angles are significant factors, but it was approaching significance for the load effect ($p = 0.091$). When testing only for the standing posture, the curvature became more kyphotic at higher load and higher flexion angle ($p<0.000$). The increased trunk flexion from 0° to 30° and knee flexion decrease from 180 to 45 (standing to squatting posture), caused the lumbar curvature to change from lordosis to kyphosis (Table 1). Also, using post hoc test the results show the lumbar curvature significantly changed to kyphosis in the trunk flexion from neutral to 15° to 30° flexion ($p<0.05$).

SPINAL INTERNAL LOADING

Table 2 shows the ANOVA results for the effects of load, knee and torso flexion effect and their interaction on external moment, compressive and shear force ($p<0.05$). All static tasks were sagittally symmetric here; hence the value of lateral shear was computed to be zero and only anterior-posterior shear was analyzed in this study. The results of ANOVA for trunk position showed a significant effect ($p = 0.00$). Based on multiple comparison (Tukey tests), this is especially true when comparing neutral to 15° to 30° trunk flexion, the external moment, compression force and shear force showed significantly increase ($p = 0.00$).

The ANOVA results for external load showed significant effect ($p = 0.000$). Post hoc testes have showed with increasing external load from 0 to 10 to 20 kg, the external moment showed significantly increase ($p<0.05$), but the significance effect for compression force was between 0 and 20 kg and 10 and 20 kg, for shear force was only between 10 and 20 kg ($p<0.05$).

Also, external moment, compression and shear force were significant due knee effect ($p = 0.00$). As the knee flexion decrease (i.e., the squatting posture was assumed), the external moment and compression force showed significantly increase but shear force significantly decreased.

DISCUSSION

The objective of this study was to determine the lumbar curvature and spinal loading for different trunk and knee posture and external loads during static holding tasks. Such postures and loads are common in many activities such as athletic and manual material handling tasks and have been recognized as a risk factor for back injuries (Arjmand and Shirazi Adl, 2006).

Lumbar curvature: Changes in the curvature during movement influence the spine stresses. During various occupational daily activities, the sagittal curvature of the human lumbar spine (i.e., lordosis) changes, influencing the mechanics of the entire human spine (Shirazi Adl and Parnianpour, 1999; McGill *et al*., 2000; Mitniski *et al*., 1998). It affects the distribution of the external loads among passive and active systems that may alter the equilibrium and stability conditions of the structure as well as the state of stress in the lumbar segmental components. In erect postures, the lumbar lordosis is reported to changes as external loads are added through weights carried in hands (Shirazi Adl *et al*., 2002).

Although squat lifting is generally considered to be a safer lift than the stoop lift, the advantages in maintain lordosis during lifting tasks are less well understood. A kyphotic lift is recommended as it utilizes the posterior ligament more, while in contrast, an increase in erector spinae activities is considered beneficial in lordoric and straight-back posture (Shirazi Adl and Parnianpour, 1999; Shirazi Adl *et al*., 2002).

The biomechanical literature regarding the advantage of maintaining or flattening the lumbar curvature during moderate lifting is fragmented and inconclusive, as Cholewicki and McGill observed that the experienced power lifters performed their lifts with relatively small lumbar flexion angle (Cholewicki and McGill, 1992).

In this study, with increasing external load from 0-10, 10-20 kg, the lordosis changes to kyphosis but this difference was not significant (Table 1 and 2; $p = 0.091$). When tasks in standing posture (excluding squatting) were examined, the external load effect on the lumbar curvature became significant, as reported in the previous reports (Anderson *et al*., 1986; Arjmand and Shirazi Adl, 2006; David and Whittle *et al*., 1996; Shirazi Adl and Parnianpour, 1999). Slight flattening of the lumbar spine under large compression loads reduces the maximum disc fiber strains and required equilibrating moments without adversely affecting the disc pressure and ligament forces (El-Rich *et al.*, 2004; Shirazi-Adl and Parnianpour, 1999; Shirazi-Adl *et al.*, 2002). Recently Shirazi-Adl *et al.* (2005) have shown minimal changes in posture (posterior pelvic tilt and lumbar flattening) substantially influenced muscle force, internal loads and stability margin.

Although literature review related to the effect of knee position on lumbar curvature is fewer in numbers, these studies have revealed the role of knee position on the lumbar posture (Chaffin and Andersson, 1999; Eklund and Liew, 1991; Farfan, 1995; Lee, 2004; Shin *et al.*, 2004; Akira and Tatsuro, 1995). Eklund and Liew (1991) have shown, hip and knee angles are important determinant for lumbar posture, but hip has higher effect than knee flexion in changing the lumbar curvature. Most of studies were evaluated the lumbar posture only while standing posture (knee flexion 180°) and some others like Akira and Tatsuro (1990) have studied the role of knee position without considering the load effect.

When knee angle decreases (subject squats), pelvis is rotated backward and the lumbar rotation is already present before the torso leaves the erect position or the person holding the weights (Chaffin and Andersson, 1999). The challenge to persons postural stability is much higher when holding load in squatting position. It would be expected that EMG studies may reveal higher levels of coactivation under these circumstances. Our mathematical model has the limitation that it cannot predict coactivation as we minimize the compressive load in the cost function. We expect that we are underestimating the compressive loads under these conditions.

The results of this study, also have indicated the main effect due trunk flexion on the lumbar curvature position were significant, with increasing trunk flexion, the lumbar lordosis were changed to kyphosis (Table 1). This is concomitant with previous literature (Anderson *et al.*, 1986; Shirazi-Adl and Panianpuour, 1999; Shirazi Adl *et al.*, 2002; Straker, 2003). The vast majority of spinal flexion occurs in the lumbar spine. Forward flexion from erect posture begins with flexion in the lumbar spine and it follows with rotation in pelvis. When returning to the upright posture, rotation at the pelvis occurs first and then extension of the lumbar spine predominates the pattern of rotation as the torso becomes erect. Extension of the lumbar spine begins later in the lifting sequence, as the load in the hands is increased (Shin *et al.*, 2004).

When, the subject bends forward, the posterior ligament is stretch passively and its tension is increased. Since the ligaments are viscoelastic structures this should allow them to support a greater share of the moment even at reduced amounts of lumbar flexion (Arjmand and Shirazi Adl, 2005). It was shown that the system stability was primarily provided by passive stiffness of motion segments that nonlinearly increased with higher axial compression and flexion angle. This argument is supported by the observations of more resistance to compression in flexed intervertebral joint (Bogduk and Twomy, 1992).

Some authors have believed hip flexion has higher effect than knee flexion on changing the lumbar curvature in sitting (Eklund and Liew, 1991), but some others have found a powerful relation between knee bending and decreasing of lumbar lordosis (Lee and Chen, 2000; Akira and Tatsuro, 1995). Chaffin and Andersson (1999) have concluded that the pelvis is rotated backwards as person squats, an increasing degree of pelvic contribution to trunk rotation is already present before the torso leaves the erect position. It could be theorized that hamstring tightness play an increasingly significant role in restraining pelvic rotation as the knee goes into deep bending position.

Internal loads: Results of past epidemiologic studies have implicated trunk combined or simultaneously occurring motions and loading as potential risk factors for occupation-related low back disorders and have emphasized the importance of quantifying three-dimensional forces to understand better the loads on the spine during work (Fathallah and Narras, 1998).

The results of this study have shown that the external moment and internal loads have significant changes due to all independent variables: load, knee and trunk posture (Table 1 and 2) in accordance with previous studies (Burgress-Limerick, 2002; Dolan *et al.*, 1994; Straker, 2003; Van Dieen *et al.*, 1999). The predicted loads are much lower than the strength of the endplate or risk of injuries to posterior elements of vertebrae. Many researchers have concluded the spinal load level that approach spine tolerance limits 3400 N for compression (NIOSH, 1997) and 1000 N for shear (McGill, 1996) to be related to higher incidence rates of LBD.

In this study, the results of the ANOVA for shear force showed a significant three way interactive effect $T \times K \times L$ ($p<0.01$). The maximum shear force predicted to be 158.8 N on standing posture with 30° trunk flexion and holding 20 kg external load. In Van Dieen's study, shear forces on lumbar spinal motion segments with holding 15 kg and two type knee positions were reported to be about 450 N (Van Dieen *et al.*, 1999). The results of ANOVA have indicated that there were significant effect of knee and trunk flexion angle and interactive effect between them ($T \times K$) in related internal loads and external moment (Table 2).

In the other hand with increasing trunk flexion from neutral to 15 to 30° in knee straight(standing posture) all variables: external moment compression and A-P shear force were increased but with decreasing knee flexion angle (squat position) from 180 to 45°, the external moment and compression force were increased, but the A-P shear force were decreased (Table 1). Trunk flexion caused the increasing compression and shear forces due to weight of upper trunk; in addition the weight of load ultimately would increase LBP risk (Bolte *et al*., 1998; Burgress-Limeck, 2003; Marrass, 1991; Van Dieen *et al*., 1999), although in such posture (stoop), compression force were lower than the squat posture (Dolan *et al*., 1994). Squat technique also may cause excessive back load due to decreasing trunk extensor strength and lifting capacity (Burgress-Limeck, 2003) and balance loss (Straker, 2003).

The results have shown, there were a significant interaction between load and trunk position L×T, knee and load L×K as a function of shear force (Table 2) ($p<0.05$). The high value of external moment or compression force was in the tasks that knee and trunk were flexed (knee flexion 45° and trunk flexion 30°). On the other hand, the high value of compression force and external load due to increased Erector Spinae muscle activity. In this situation the pelvis is in posterior tilt and muscle forces have vertical direction and they are more compressive type but the high value of shear force was in tasks with combination trunk flexion (30°) and knee flexion 180° (straight) due to increasing of anterior pelvic tilt and increasing shear component of reactive joint forces. In conclusion, posterior pelvic tilt in squat posture has the more decreasing role for shear force than the increasing role for external moment.

Table 1 have shown in squat tasks, the value of the external moment and internal loads in 15° trunk flexion are very near to neutral tasks. This is probably due to a little trunk flexion (at mostly equal to 15°) in neutral posture. In this study the mean of trunk bending angle in tasks neutral posture were 12.6°, which it was near to value 15° trunk flexion.

The results of ANOVA have shown the interaction L×K is significant in related the shear force ($p<0.002$) that is, the most shear force was in tasks with holding 20 kg load in straight knee postures,. When the external load is increased, the external moment and reaction force of knee joint also increased, so the shear force would increased due to trunk bending in this situation. Although in squat tasks with high value of external load, the shear force was increased but it is lower than stoop tasks. It could be theorized that posterior pelvic tilt has a significant role in decreasing the shear force component and increasing the vertical component. Anderson *et al*. (1986) have concluded that hamstring tightness play significant increasing role in restraining pelvic rotation as the knee bends. The maximum shear force in this study was 158.8 N that was in task with trunk bending, knee straight, holding 20 kg, while the compression force in those tasks were 1295.5 N. The most compression force was 1614 N in task that both trunk and knee were flexed (30° and squat, respectively) while holding 20 kg weight.

So the squat tasks with or without the external load, would decrease the shear force, but such situation is not recommended for lifting especially when it is concomitant with trunk bending. Consequently lumbar spinal motion segments fail under compression at levels in between about 2 to 10 KN, although their strength appears to be age and sex dependent., but it is for shear forces were reported 1 to 2.5 KN, so the shear force during lifting do not pose a serious risk (Van Dieen *et al*., 1999). In conclusion the balance loss during lifting or holding load, which may cause excessive back load, is more likely to occur when using the squat technique as compared to the stoop techniques.

CONCLUSIONS

- Effects of external load, trunk and knee flexion position on lumbar curvature and spinal internal loads during static holding tasks were investigated using both *in vivo* experimental and biomechanical model studies.
- Change of lumbar curvature towards kyphotic posture at higher loads became significant in standing posture. The stability and loss of balance in squatting posture was noted in this study and some subjects had difficulty performing these tasks.
- Decrease in lumbar extension while lifting external load is the supportive role against compression force and to prevent instability of motion segments.
- Decrease in lumbar extension during knee squatting is due to posterior pelvic tilt and due to tightness of hamstring.
- External moment and internal loads (compression and shear forces) changed due to external load, trunk and knee flexion angle effects.
- Highest value of external moment and compression force was predicted in tasks that both trunk and knee were flexed, but in these tasks shear force was less due to pelvic rotation.
- In spite of less shear force in squatting, this technique is not safe or recommended for lumbar spine especially when it would be with high levels of trunk flexion and heavy weight.

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