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Lumbar Extension during Stoop Lifting is Delayed by the Load and Hamstring Tightness

Risa Iwasaki, PT[1], Ginga Yokoyama, PT[2], Satoshi Kawabata, PT[2], Tomotaka Suzuki, PT, MS[3, 4]*

[1] *Department of Rehabilitation, Saiseikai Kanagawa Hospital, Japan*
[2] *Department of Rehabilitation, Ushioda General Hospital, Japan*
[3] *School of Rehabilitation, Faculty of Health and Social Services, Kanagawa University of Human Services: 1-10-1 Heisei-cho, Yokosuka, Kanagawa 238-8522, Japan*
[4] *Unit of Rehabilitation Sciences, Nagasaki University Graduate School of Biomedical Sciences*

Abstract. [Purpose] This study investigated the relationship between lumbar pelvic rhythm and the physical characteristics of stoop lifting. [Subjects and Methods] Participants performed a stoop lifting task under two conditions: with and without load. We assessed the lumbar kyphosis and sacral inclination angles using the SpinalMouse® system, as well as hamstring flexibility. During stoop lifting, surface electromyograms and the lumbar and sacral motions were recorded using a multi-channel telemetry system and flexible electrogoniometers. [Results] In the initial phase of lifting, lumbar extension was delayed by load; the delay showed a negative correlation with sacral inclination angle at trunk flexion, whereas a positive correlation was observed with electromyogram activity of the lumbar multifidus. Additionally, a positive correlation was observed between sacral inclination angle and hip flexion range of motion during the straight leg raise test. [Conclusion] We found that a disorder of the lumbar pelvic rhythm can be caused by both load and hamstring tightness. In the initial phase of stoop lifting, delayed lumbar extension is likely to lead to an increase in spinal instability and stress on the posterior ligamentous system. This mechanism shows that stoop lifting of a load may be harmful to the lower back of people with hamstring tightness.

Key words: Lifting, Lumbar pelvic rhythm, Hamstring flexibility

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INTRODUCTION

Low back pain is a common disorder that often affects sufferers' daily performance[1]. In recent years, the psychological aspects of low back pain have gained much attention, although specific physical characteristics and body movements largely influence the etiology of this disorder. Effective preventive measures against low back pain are thus desired.

The factors contributing to the development of low back pain include frequent lifting. Two types of lifting movements have been associated with low back pain: *stoop lifting*, which involves extension of the trunk and hip and knee joints; and *squat lifting*, which requires maximal flexion of the knees. These movements differ in terms of the burden placed on the back muscles during lifting; however, in terms of prevention of low back pain, no apparent differences have been observed[2]. Bazrgari et al.[3] recently proposed that squat lifting was safer because of the smaller trunk extension torque requirement; however, supporting evidence for this theory is insufficient, and the risks posed by stoop lifting remain unclear.

Nelson et al.[4] reported that lumbar and pelvic motions occur sequentially during stoop lifting. Lumbar pelvic rhythm reduces the load on the lumbar extensor muscles, the underlying intervertebral joints, and the intervertebral discs; these effects might protect the lumbar region against high levels of stress[5]. Studies of low back pain[6, 7], load[8], lifting velocity[9], and posture[10] have reported on lumbar pelvic rhythm, although findings pertaining to the prevention of low back pain are limited.

Physical characteristics such as the alignment of the spine and the flexibility of the trunk and lower extremities are important for establishing the mechanism of the development of low back pain. The consideration of physical characteristics is also necessary for analyzing the mechanism of lumbar pelvic movement during stoop lifting. However, few such studies have been performed and previous studies have mainly focused on the effects of the act of lifting. Therefore, the purpose of this study was to elucidate the relationship between physical characteristics and the lumbar pelvic rhythm during stoop lifting. Furthermore, based on the analysis of muscle activity, we investigated the changes of the lifting strategy caused by physical characteristics and variations in load. Low adaptability resulting from unfavorable physical characteristics is presumed to greatly enhance the effects of the load.

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*Corresponding author. Tomotaka Suzuki (e-mail: suzuki-tm@kuhs.ac.jp)

SUBJECTS AND METHODS

The participants were 26 students (16 men and 10 women; mean age ± standard deviation [SD]: 21.8 ± 3.8 years) attending Kanagawa University of Human Services, Yokosuka, Japan. None of the participants had a history of physical functional impairment, which may have rendered stoop lifting difficult. Additionally, participants were ascertained to have no current low back pain. Participants provided their informed consent before taking part in the experiment. This study was performed with the approval of the Research Ethics Committee of Kanagawa University of Human Services (24–044).

Participants performed stoop lifting, which involved extending the elbows and knees while bending the trunk. Participants stood with their feet shoulder-width apart. They were instructed to grasp the handles of a plastic box (28 × 40 × 25 cm; weight, 600 g) placed 10 cm in front of their toes and to lift it by trunk extension. Stoop lifting was performed under two conditions: with load (20% of the participant's body weight) and without load. The load was symmetrically set in the box. Participants practiced stoop lifting prior to the actual experiment to ensure that the lifting task was correctly performed. Each stoop lifting condition was performed five times, with lifting with and without load performed alternately.

We assessed the lumbar and sacral postures and the flexibility of the spinal column and the lower extremity before the lifting task. The SpinalMouse® system (Idiag, Volkerswill, Switzerland) was used to measure standing sagittal curvature[11]. This measurement has been validated for the evaluation of global lumbar mobility during trunk flexion[12]. The SpinalMouse® device was guided paravertebrally along the spinal column from C7 to S3. The measurements were conducted under two conditions: standing upright and at the starting position of stoop lifting (trunk flexion). We analyzed lumbar curvature (T12-L1 to the sacrum) and the sacral angle to assess lumbar and sacral postures. Kyphotic angles are expressed as positive values.

Surface electromyograms (EMG) were recorded using active silver/silver chloride surface electrodes (ZB-150H; Nihon Kohden Corp., Tokyo, Japan) and a multi-channel telemetry system (WEB-1000; Nihon Kohden Corp., Tokyo, Japan). The distance between the electrodes of the electrode telemeter was 10 mm. After cleaning the skin with alcohol, the electrodes were placed over the belly of the right semitendinosus (ST), external oblique (EO), internal oblique (IO), rectus abdominis (RA), latissimus dorsi (LD), and erector spinae (ES) at the Th9 and L1 levels, and the lumbar multifidus (LM) at L5/S1. The raw signal was amplified, bandpass filtered (5–500 Hz), digitized at 1,000 Hz, and stored for off-line analysis on the laboratory computer.

We used two flexible electrogoniometers (SG150; Biometrics Ltd., Gwent, UK) to analyze the lumbar and sacral motions in the sagittal plane. In the lower back region, the proximal electrogoniometer end block was placed at the L1 level, and the distal end block was placed at the L5 level. The proximal end block of another electrogoniometer was placed at the S1 level of the sacral region of the back, and the distal end block was fixed to the outside object. Using the telemetry system previously described, the angle and EMG data were synchronized, and wirelessly stored on a laptop computer.

We measured hip flexion range of motion using the passive straight leg raise (SLR) test and evaluated hamstring flexibility. Additionally, we measured finger floor distance (FFD), which reflects the flexibility of the trunk and hamstrings.

In this study, the analysis range during stoop lifting was determined as the point where backward rotation of the sacrum began to the point where extension of the lumbar spine. The starting point of each participant was defined as 0% and the end point was defined as 100%. This range was equally divided into 10 phases; the change in extension angle in each phase was normalized to the total extension of all the phases.

To normalize the duration of each trial, the angle data were resampled using bio-information analysis software (BIMUTAS II; Kissei Comtec, Matsumoto, Japan) and the arithmetic mean of each phase was calculated. After the EMG signal was full-wave rectified and resampled, mean EMG activities were calculated for each phase. These values were then normalized to each subject's maximum voluntary contraction (MVC), which was measured before the lifting task; the measurement position was based on McGill's method[13], and Daniels and Worthingham's Muscle Testing[14].

All the statistical analyses were conducted using IBM SPSS statistics 17 for Windows (SPSS Inc., Chicago, IL, USA). For EMG activity, the main effects of the load and the lifting phase were compared using two-way repeated-measures analysis of variance (ANOVA). In addition, the simple main effect of the load was examined using post hoc t-tests. Pearson's product moment correlation coefficient was used to investigate the relationship among physical characteristics (i.e., lumbar kyphosis angle, sacral inclination angle,SLR, and FFD). Statistical significance was accepted for values of $p < 0.05$ in all tests.

RESULTS

During stoop lifting without load, changes involving the posterior tilt of the pelvis were greater during the first half of the lifting movement, whereas changes involving the extension of the lumbar spine were greater during the latter half (Fig. 1). When the load was applied, the extension of the lumbar spine, which is supposed to occur after initiation of the movement, was delayed and the corresponding graph showed a shift to the right. On the other hand, the pelvis showed no marked angular changes after the initiation of the movement.

In two-way repeated measures ANOVA, a significant main effect of load was found in EMG activities of all the target muscles: the ST, EO, IO, RA, LD, ES (Th9), ES (L1), and LM muscles (respectively: $p < 0.001$, $p = 0.004$, $p < 0.001$, $p = 0.019$, $p < 0.001$, $p < 0.001$, $p < 0.001$, $p < 0.001$). Furthermore, there were significant interactions between the load and the lifting phase in the ST, RA, LD, ES (Th9), ES (L1), and LM muscles (respectively: $p < 0.001$, $p = 0.020$, $p < 0.001$, $p < 0.001$, $p < 0.001$, $p < 0.001$).

In the ST, EO, IO, LD, ES (Th9), and ES (L1) muscles, post hoc paired t-tests showed that the application of load resulted in a significant increase in EMG activity in all phases (Table 1). For the RA muscle, the application of load resulted in a significant increase in EMG activity in the 0–10% phase. In the LM muscle, EMG activity showed no difference due to load in the 10–20% phase; however, in the other phases, its EMG activity increased significantly. In terms of physical characteristics, a negative correlation was observed between the sacral inclination angle and the lumbar kyphosis angle at trunk flexion; a positive correlation was detected between sacral inclination angle and SLR, whereas no correlation was observed with FFD (Table 2).

Because the load caused a delay in lumbar extension in the initial phase of stoop lifting, we calculated the difference in lumbar extension in relation to load in each phase: [delayed lumbar extension = normalized extension without load − normalized extension with a load], and analyzed the relationship between physical characteristics and differences in EMG activities in relation to load. The results show a negative correlation between delayed lumbar extension and the sacral inclination angle at trunk flexion in the 10–20% phase, and a positive correlation between delayed lumbar extension and EMG activity of the LM in the 0–10% phase (Table 3).

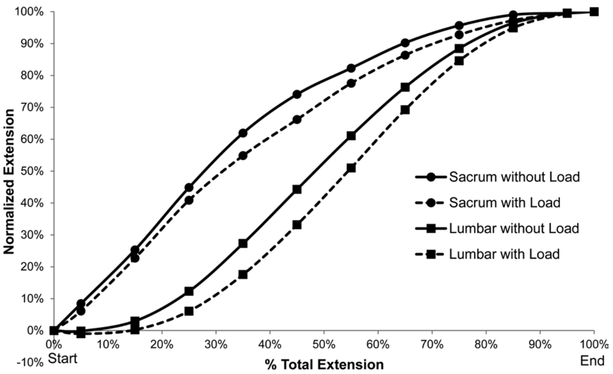

Fig. 1. The lumbar pelvic rhythm during stoop lifting with or without load
Normalized extension in lumbar spine and sacrum are plotted against the percentage of total extension. The graph of lumbar spine is shifted to the right because of a delay in lumbar extension at the initial phase of lifting.

Table 1. Mean EMG activities in each phase during stoop lifting with or without load

Muscle	EMG activities in each phase									
	0–10%	10–20%	20–30%	30–40%	40–50%	50–60%	60–70%	70–80%	80–90%	90–100%
Semitendinosus										
Without Load	9.4 (6.0)	9.3 (6.5)	8.6 (6.1)	7.7 (5.1)	6.7 (5.0)	5.5 (4.6)	4.3 (3.9)	3.4 (3.7)	2.7 (3.0)	2.5 (2.9)
With Load	16** (12)	15** (11)	14** (8.7)	12** (6.2)	9.9** (5.9)	8.1** (4.9)	6.8** (4.8)	6.0** (4.6)	5.5** (4.2)	5.0** (3.9)
External oblique										
Without Load	2.1 (1.7)	2.1 (1.8)	2.2 (1.7)	2.2 (1.7)	2.2 (1.7)	2.2 (1.7)	2.2 (1.7)	2.2 (1.7)	2.2 (1.7)	2.2 (1.7)
With Load	3.3** (3.4)	3.1** (2.8)	3.6* (3.9)	3.4** (3.1)	3.6* (4.0)	3.2** (2.9)	3.1** (2.6)	3.0** (2.4)	3.2* (3.8)	3.1* (3.5)
Internal oblique										
Without Load	3.0 (4.1)	3.5 (5.2)	4.2 (5.8)	5.1 (6.5)	5.3 (6.8)	5.9 (7.5)	6.4 (8.6)	6.7 (9.4)	6.3 (8.6)	6.0 (8.2)
With Load	5.9** (5.9)	6.6** (7.7)	7.2** (8.0)	7.9** (8.2)	8.8** (8.8)	9.4** (9.2)	9.2** (9.6)	10** (11)	9.2** (11)	9.0** (9.8)
Rectus abdominis										
Without Load	1.4 (1.3)	1.4 (1.3)	1.4 (1.3)	1.4 (1.3)	1.4 (1.4)	1.4 (1.4)	1.4 (1.3)	1.4 (1.3)	1.4 (1.3)	1.4 (1.3)
With Load	1.7* (1.5)	1.7 (1.5)	1.6 (1.5)	1.5 (1.4)	1.5 (1.3)	1.5 (1.4)	1.4 (1.3)	1.4 (1.3)	1.4 (1.4)	1.4 (1.4)
Latissimus dorsi										
Without Load	4.3 (4.6)	4.7 (4.4)	4.3 (3.7)	4.2 (3.3)	4.1 (3.1)	3.8 (2.7)	3.4 (2.3)	2.8 (1.9)	2.3 (1.6)	2.2 (1.6)
With Load	11** (10)	10** (8.3)	9.3** (6.3)	8.6** (6.2)	7.4** (4.6)	6.7** (4.1)	6.3** (4.2)	5.3** (3.5)	4.4** (2.9)	4.0** (2.4)
Erector spinae (Th9)										
Without Load	7.5 (4.8)	7.1 (4.0)	6.9 (3.7)	6.4 (3.6)	6.4 (3.5)	6.2 (3.6)	5.5 (3.2)	4.7 (2.7)	4.0 (2.5)	3.7 (2.1)
With Load	21** (10)	22** (9.9)	19** (7.7)	16** (6.6)	14** (5.9)	13** (6.1)	12** (5.5)	10** (4.6)	9.0** (4.0)	8.3** (3.6)
Erector spinae (L1)										
Without Load	7.5 (5.1)	11 (6.0)	13 (6.2)	15 (6.4)	15 (6.2)	14 (5.9)	12 (5.4)	9.4 (4.6)	7.3 (3.5)	6.7 (3.7)
With Load	13** (8.7)	16** (9.6)	19** (9.0)	21** (9.6)	23** (12)	24** (13)	22** (13)	18** (10)	15** (7.7)	12** (6.1)
Lumbar multifidus										
Without Load	8.0 (4.4)	12 (4.6)	14 (4.4)	14 (3.8)	14 (3.3)	13 (3.9)	11 (2.9)	8.2 (3.2)	6.6 (2.9)	5.7 (2.3)
With Load	11* (8.2)	14 (9.4)	18** (8.3)	21** (9.2)	20** (8.5)	19** (8.3)	17** (6.9)	15** (6.2)	12** (5.3)	9.9** (3.7)

EMG = electromyogram. EMG activities are normalized to each subject's maximum voluntary contraction (%MVC). Data are expressed as means (standard deviation).
*$p < 0.05$ compared with "Without Load". **$p < 0.01$.

Table 2. Summary of physical characteristics and their relationship with the sacral inclination angle

Physical characteristics	M (SD)	Sacral inclination angle at trunk flexion Pearson's r
Upright		
lumbar kyphosis angle	−27.6 (10.5)°	-
sacral inclination angle	17.0 (8.3)°	-
Trunk flexion		
lumbar kyphosis angle	31.4 (8.7)°	−0.72**
sacral inclination angle	66.5 (8.4)°	-
Hip flexion ROM during SLR test	78.3 (11.0)°	0.53**
Finger floor distance	6.0 (9.8) cm	0.31

ROM = range of motion; SLR = straight leg raise
**p < 0.01

Table 3. Relationship among differences in lumbar extension and EMG activities of the lumbar multifidus, with or without load, and sacral inclination angle during the initial phase of stoop lifting

	Delayed lumbar extension Pearson's r		
	0–10%	10–20%	20–30%
EMG activities of Lumbar multifidus	0.45*	0.20	0.14
Sacral inclination angle at trunk flexion	−0.35	−0.45*	−0.39

EMG = electromyogram
*p < 0.05

DISCUSSION

The lumbar and pelvic movements caused by stoop lifting were similar to those reported by a previous study4). However, the lumbar pelvic rhythm varied depending on the presence of the load. In addition, the load-induced increase in muscle activity also varied based on the muscle and timing of movement.

All components of the spinal column (i.e., intervertebral disc, spinal ligaments, and facet joints) contribute to spinal stability. Moreover, the spinal muscles provide mechanical stability to the spinal column like guy wires15). Normal lumbar pelvic rhythm protects these tissues against high levels of stress5). The delay in lumbar extension shifts the extensor torque demand to the powerful hip extensors, and the lumbar extensor muscles are activated after the trunk is raised, minimizing the external moment arm16). Our present study also showed that in the initial phase of lifting, the extension of the lumbar spine was delayed compared to that of the pelvis and the ST muscle was active. These features were increased by load.

Similar to the ST muscle, the application of a load also induced a marked increase in EMG activity in the LD and ES (Th9) muscles during the initial phase of lifting. When a heavy load is lifted, the delay in lumbar extension might cause the lifting movement to use a larger number of upper back muscles, indicating the strategies used by the thoracic and the lumbar spines during lifting with load are different from each other.

During lifting, we consider the EO and IO muscles acted to increase the intra-abdominal pressure. Therefore, we recommend using the Valsalva method to increase the intra-abdominal pressure in order to reduce the compressive force exerted on the spinal column. In addition, the contractile forces exerted by the EO and TA muscles are posteriorly transmitted to the thoracolumbar fascia, providing stability in the lumbar region like a corset5). In the present study, the RA muscle activity relatively showed less of this type of effects, and thus its involvement in lifting may be minimal.

The ES (L1) and LM muscles displayed smaller amounts of muscle activity in the initial and latter phases of the lifting movement. In this study, no difference due to load was found in the EMG activity of the LM muscle in the 10–20% phase, and the phase showing the maximum amount of activity of the ES (L1) muscle was delayed after load application. This observation may be related to the load-induced delay of lumbar extension.

During lifting, the peak extensor moment is generated in the initial phase of lifting (approximately 5–20%), and passive tissues assist the back muscles through passive tension17). If the lower back muscles activate less with delayed lumbar extension, then stress on the posterior ligamentous system increases. In addition, this stress would recruit a compensatory low back muscle force, which might affect ligamento-muscular synergy18). Consequently, a disorder of lumbar pelvic rhythm and low back pain may be caused, as has been reported for persons with a history of low back pain, who show greater lumbar motion during the initial phase of trunk extension7).

Analysis of the relationships among the prolongation of delay in lumbar extension, physical characteristics, and EMG activity has revealed that a lower sacral inclination angle at trunk flexion was associated with a greater delay in lumbar extension during the initial stages of lifting with load than that without load. In addition, analysis of the relationships among sacral inclination angle at trunk flexion, lumbar kyphosis angle at trunk flexion, and SLR showed that a decrease in sacral inclination angle at trunk flexion is attributable to a decrease in hamstring flexibility, causing compensatory hyperflexion of the lumbar spine. Furthermore, the results of this study show that the delay in lumbar extension is also related to the EMG activity of the LM muscle during the initial phase of lifting.

Hamstring length influences the pelvic angle in the toe-touch position19). McClure et al.7) reported that hamstring length was not correlated with the kinematic characteristics during the return to an upright position from a bending position, whereas Esola et al.6) reported that hamstring flexibility was correlated to the motion of subjects with a history of low back pain during forward bending.

This study on stoop lifting revealed the effect of hamstring muscle flexibility on the lumbar pelvic rhythm due to the application of a load. The findings show the potential harmful effects of stoop lifting, which were previously unclear, and the impact of load application and the flexibility of the hamstring muscles. Application of a load affects the lumbar pelvic rhythm by causing a delay in lumbar extension. Subjects with less flexible hamstring muscles are unable to achieve a sufficient inclination of the pelvis, and thus they adopt a strategy aimed at delaying the lumbar extension even further by performing a compensatory hyperflexion of the lumbar spine. Therefore, the initial phase of lifting causes an extension of the period of instability of the spinal column, as well as increasing the stress on the posterior ligamentous system, which must subsequently be followed by rapid lumbar extension. Additionally, vigorous contraction of the back muscles with the lumbar spine maximally flexed may damage the intervertebral discs5). The phenomena described in this study may have harmful effects on the lumbar region and are likely to cause low back pain.

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