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Stooping and squatting postures are seen in a number of industries (e.g., agriculture, construction) where workers must work near ground level for extended periods of time. The focus of the current research was to evaluate a knee support device designed to reduce the biomechanical loading of these postures. Ten participants performed a series of sudden loading tasks while in a semisquat posture under two conditions of knee support (no support and fully supported) and two conditions of torso flexion (45 and 60°). A weight was released into the hands of the participants who then came to steady state while maintaining the designated posture. As they performed this task, the EMG responses of the trunk extensors (multifidus and erector spinae) were collected, both during the "sudden loading" phase of the trial as well as the steady weight-holding phase of the trial. As expected, the effects of torso flexion angle showed significant decreases in the activation of the multifidus muscles with greater torso angle (indicating the initiation of the flexion–relaxation response). Interestingly, the results showed that the knee support device had no effect on the activation levels of the sampled muscles, indicating that the loss of the degree of freedom from the ankle joint during the knee support condition had no impact on trunk extensor muscle response. The a priori concern with regard to these supports was that they would tend to focus loading on the low back and therefore would not serve as a potential ergonomic solution for these stooping/semisquatting tasks. Because the results of this study did not support this concern, further development of such an intervention is underway.

Key Words: lumbar, EMG, kneeling

Our ergonomic intervention effectiveness research has led us into the agricultural environment, where static stooping, semisquatting, squatting, and/or kneeling postures are common as workers harvest produce. During these tasks, agriculture workers can be seen shifting from stooped postures (straight knees and full flexion of the torso) to squat postures (full knee flexion and upright torso postures) to kneeling postures (knees on the ground and upright torso postures) and a variety of postures in between. The workers often alternate between these positions in order to gain relief from the discomfort of the static muscle exertions in various regions of the body. A generic ergonomic intervention strategy would seek to lower the center of mass of the torso without relying on the squatting posture (which requires significant knee extensor muscle exertion), a stooped posture (which places high loads on the low back—both active and passive tissues), or a full kneeling posture (which has been shown to be a contributor to knee complaints through mechanical trauma).

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A study by Dieen et al. (1997) evaluated one such intervention in radish harvesting. These authors noted that this particular task requires the hands of the workers to be almost at ground level. When this task is done in a kneeling posture, the workers use a full knee flexion and "the back flexed during more than 70% of the total working time" (p. 355). The intervention these authors evaluated was a chair that included a seat pan and seat back that rested directly on the ground and allowed for a more upright, but twisted, torso posture. This study involved the evaluation of biomechanical measures of effectiveness including EMG, spinal shrinkage, and extensor muscle force, and the study included ratings of subjective discomfort and ratings of perceived exertion. These authors showed positive biomechanical effects through reduced normalized activity of the longissimus muscles and reductions in both the discomfort and rating of perceived exertion as workers performed the radish-harvesting task using the intervention.

In a pepper-harvesting task, workers are asked to harvest the produce anywhere from 0.2 m to 1.5 m from the ground. This position limits the utility of the fully seated posture because it would require unacceptably high levels of shoulder flexion and/or abduction. This required consideration of a semi-squatting intervention, using such a structure on which the worker could kneel and thereby lower the torso to the appropriate working posture. One solution under consideration was the development of a mobile kneeling bench that the worker could place on the ground and kneel upon while harvesting, providing a semikneeling posture. This bench would eliminate the full flexion of the knees and reduce (but not completely eliminate) the amount of trunk flexion. The impact of this posture on low back biomechanics was uncertain, and this uncertainty led to the development of the current study.

One component of the scientific literature relevant to this topic is focused on full kneeling postures. Much of the important research in this area has been motivated by challenges in the mining industry (Gallagher et al., 1994, 1997, 2002). In a study of the effects of the kneeling posture on extension strength and maximal muscle activation levels, Gallagher (1997) showed that the kneeling posture generated a 15% reduction in peak trunk extension torque as compared with the standing posture. This implies that for a given trunk extension torque requirement, the kneeling worker is required to exert a greater percentage of their capacity than would a standing worker in a similar trunk flexion posture. In another study of the biomechanics of the full kneeling posture, Gallagher et al. (1994) showed significant differences in muscle activation levels (and subsequent estimates of spine compression) during standardized lifting tasks in kneeling and stooped postures. In this study, participants lifted three loads (15 kg, 20 kg, or 25 kg) from the ground to shelves at two heights (0.35 m or 0.7 m) and in two asymmetric postures (sagittally symmetric or an 90° asymmetric lifting motion). Lifting tasks in all combinations of these variables were performed using both a stooped posture and a kneeling posture and muscle activity levels were captured using electromyography. The results of the study showed significantly higher erector spinae muscle activities in the kneeling postures as compared with the stooped postures. When an EMG-assisted model was used, the results showed much higher spinal compression in the kneeling posture and higher anterior–posterior shear forces in the kneeling posture during asymmetric lifting. The authors attribute some of the increased activation of the erector spinae muscles to the increased reliance on these muscles because in the kneeling postures the potential contribution of the gluteal and other hip extensors is reduced.

This description of the underlying biomechanics of these kneeling postures highlights several biomechanical considerations. First, this differentiation between various trunk extensor muscle groups as a function of stooped vs. kneeling posture leads one to wonder whether there is a potential further differentiation within the erector group itself. Even in the same extensor group, the function of the muscles could be different. Bergmark (1989) hypothesized that the function of the multifidus muscle group is to transfer forces and to control lordosis while the iliocostalis and longissimus extend the spine. It is possible that these two functions may be differentially affected by the posture of the lower extremity and the relative contribution of the other extensor muscles (gluteal and hip extensors). Second, what might be the biomechanical response of low back musculature in the case where the kneeling posture is not complete (i.e., the knees are not on the ground) but instead are supported by a knee support similar to concept that was used in the Balans chair design (Lander et al., 1987)? The interesting factor in this design is that the degree of freedom of the ankle joint is still lost in this scenario (i.e., the lowest point of freedom is the knee joint), but the angle of the knee joint is considerably different than that seen in full kneeling posture and this can affect pelvic angle and thereby lumbar mechanics.

The specific aim of this work was to evaluate the differences in lumbar erector muscle activation levels while performing a task in a semikneeling posture both with and without a knee support mechanism. The task chosen for this experiment included both a sudden loading component as well as a steady-state weight-holding task.

Methods

Participants

Ten male participants were recruited for this experiment, with age, stature, and weight (and *SD*) of 27.5 (2.9) years, 176 (6.6) cm, and 73.4 (13.1) kg, respectively. None reported current or chronic low back problems. The experimental protocol was approved by the Institutional Review Board for the Protection of Human Subjects in Research. All participants provided written informed consent prior to participation.

Apparatus

A wooden structure was built to provide the lower extremity support (Figure 1). This support structure was built to be height and angle adjustable to accommodate participants of varied anthropometry. Padding was provided over the surface of the support to make the kneeling surface comfortable (the device provided full shin support, not just at the knee). Another structure was built to provide a consistent method of delivering the sudden load to the participant. This structure held the load and when the experimenter released the lever, the load was instantly transferred to the participant. This structure allowed the participant to hold onto the barbell without any hand load, so that he experienced no load to full load in an instant. The experimenter was standing on a force platform so that the instant of load delivery was registered in the data collection system.

Four pairs of surface Ag–AgCl electrodes (Model E22x, In-Vivo Metric) were used to collect the electromyographic (EMG) activity. The location

Figure 1 — The kneeling apparatus.

of the bilateral multifidus was 1.5 cm from the vertebral midline at the L4 level. The location of the bilateral longissimus was 3.5 cm from the midline of the spine at the L2 level. Although the signal collected from the two trunk extensors (longissimus and multifidus) almost certainly contained cross-talk from adjacent trunk extensor muscles (e.g., Stokes 2003), the electrode placement locations were chosen to maximize the contribution of the named muscle based on the relative cross-sectional areas of the muscles in the region (L2 to L4). An isokinetic lumbar dynamometer was used to provide the necessary static resistance for the collection of the angle-specific maximum voluntary contraction (MVC) EMG data from the lumbar extensors. This dynamometer system was also able to provide a measure of the angle-specific peak moment generated by the individual participants. This peak moment was then used to calculate the 30% of capacity, hand-held load that was used in the experimental trials.

Experimental Design

The basic design of this experiment was a randomized complete block design. The independent variables in this study were the lower extremity *support* condition with two levels (no support vs. with knee support) and torso flexion *angle* with two levels (45° vs. 60°). These two particular torso flexion angles were chosen for evaluation because of the expected changes in the relative contribution of the passive tissues to the total extensor moment (i.e., the flexion–relaxation response) as has been shown in previous research (e.g., Shin et al., 2004). The goal in the current study was to evaluate the potential interactive effects between this torso angle and knee support effects that might be generated. Each of the four combinations was repeated five times, resulting in a total of 20 trials per participant. Trial order was fully randomized.

The dependent variables in this study included a measure of the extensor muscles' response both during the sudden loading phase of the trial as well as during the steady state phase. To gather an estimate of the muscle performance, the normalized (to MVC) EMG data were calculated for each muscle in each trial, both immediately after the load was delivered (sudden loading phase) and after the subject reached steady state. Finally, because this was a sagittally symmetric task, the right and left values were averaged for each measure. The resulting dependent variables were (1) normalized EMG of the multifidus during sudden loading, (2) normalized EMG of longissimus during sudden loading, (3) normalized EMG of the multifidus at steady state, and (4) normalized EMG of longissimus at steady state.

Experimental Procedure

The experiment started with a stretching and warm-up period, and this was followed by the placement of the surface electrodes. After electrode placement, the maximum voluntary contraction of the trunk extensors was measured using a lumbar dynamometer (Mirka and Marras, 1993) while the participant assumed a sagittally symmetric, 45° forward flexion trunk angle. After completing the MVC exertions, the experimental trials began. In those trials in which the participant was not provided knee support, the experimenters used goniometers to put the participant into the correct posture (60° knee flexion and either 45° or 60° trunk flexion), as shown in Figure 2. In those trials in which the participant was provided knee support, the experimenters slid the knee support into position, such that when the participant

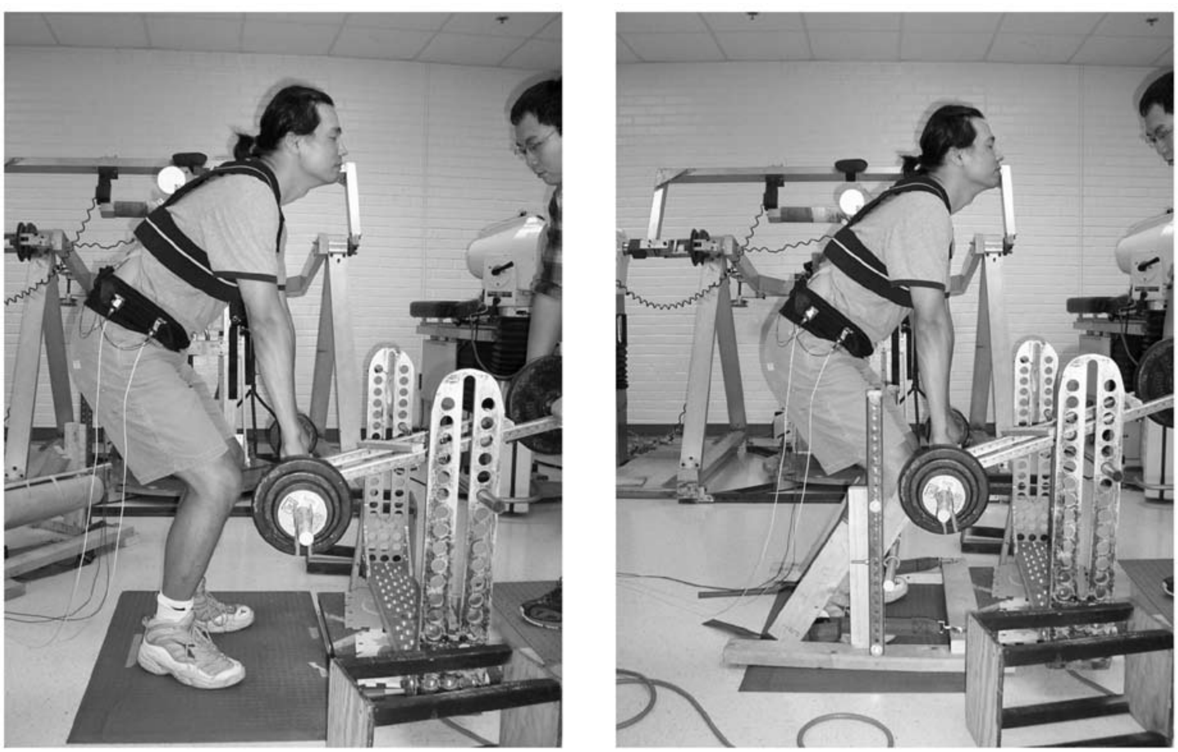

Figure 2 — A participant holding the load in free (left) and knee-supported (right) conditions.

knelt on the support the knee flexion angle was again 60° and the participant then simply bent his torso forward to achieve the required trunk flexion angle (45° or 60°), as shown in Figure 2. Once the participant was in the correct posture, the load was lifted to the participant's hands as their arms hung relaxed from their shoulders. The participant was told to grab the bar without supporting any of the weight of the load. When the back flexion and knee angles were reached, the participant was then asked to close his eyes. The closed-eyes condition was used to avoid any anticipatory or preparatory activity on the part of the participant. Within the next 10 s, the load was released and the participant had to catch and hold the load. The instructions given were to maintain the whole-body posture. Data collection lasted for 10 s (2 s before release and 8 s with load), after which time the load was taken from the participant and he returned to the upright relaxed posture. A 30-s break was provided between trials (pilot work showed this break to be sufficient to prevent cumulative fatigue for this task).

Data Processing

All the raw EMG data were filtered using a 15- to 500-Hz bandpass filter and a notch filter that eliminated 60 Hz and aliases. Once filtered, these signals were full-wave-rectified. For the MVC trials, the peak EMG value for each muscle was obtained and the greatest of these for each muscle across all MVC trials was used as the denominator in the process of normalizing the EMG data from the experimental trials. For the experimental trials, the starting point at which the load was delivered to the participant was identified using the force plate. The data collected during the next second was used to describe the sudden load response and the data collected during the 3- to 4-s interval was used to describe the steady state response. Data in these time intervals (0–1 and 3–4) were averaged and then normalized to MVC. Finally, an average of the right and left muscles of each pair was calculated.

Data Analysis

The assumptions of the ANOVA were first assessed using the graphical approach advocated by Montgomery (2001). A randomized complete block design (with subject acting as the blocking variable) was used to analyze these data. Because there were four dependent measures, a MANOVA was first conducted to assess the effect of the independent variables while controlling for the experimentwise error rate. If statistical significance of the MANOVA ($p < 0.05$ for Wilks's Λ) was found for a main effect (or interaction), then that effect (or interaction) would be further tested using univariate ANOVAs for each measure. A p value of less than 0.05 was used as the criteria for a significant effect.

Results

The MANOVA showed a significant effect for angle, but neither support nor the interaction of support and angle were found to be significant (Table 1). Subsequent univariate ANOVA revealed that angle had a statistically significant effect on the multifidus activation during both the sudden loading phase of the trial as well as the steady state response (Figure 3). These results indicate that the 60° condition was initiating the flexion–relaxation response of the multifidus muscles, as shown in previous studies (Shin et al., 2004) for both the steady state and sudden loading exertions. The lack of significance of support on these responses suggests that the loss

Table 1 MANOVA and Subsequent Univariate ANOVA Results (*F* and *p* Values)

	Support	Angle	Support × Angle
MANOVA	$F = .84$, $p = 0.12$	$F = 3.33$, $p = 0.01$	$F = 0.72$, $p = 0.58$
Sudden loading response			
Longissimus	—	$F = 0.66$, $p = 0.42$	—
Multifidus	—	$F = 5.60$, $p = 0.02$	—
Steady state response			
Longissimus	—	$F = 0.97$, $p = 0.32$	—
Multifidus	—	$F = 7.33$, $p = 0.01$	—

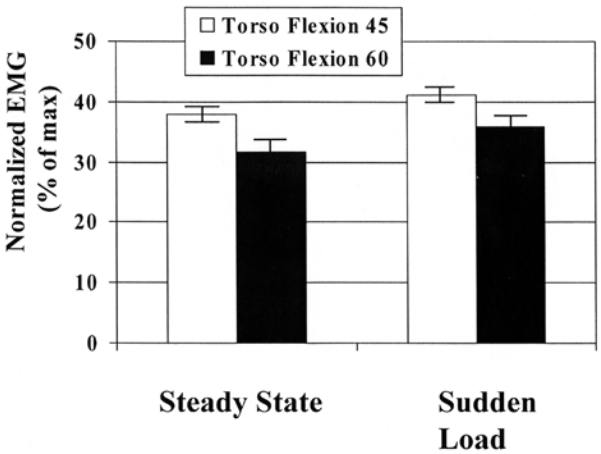

Figure 3 — Response of the normalized EMG of the multifidus to torso ANGLE conditions (both statistically significant steady state and sudden loading conditions).

of the degree of freedom of the ankle joint did not have a significant impact on the response of the low back musculature.

Discussion

In this experiment, the biomechanical responses of low back trunk muscles were studied when participants were asked to hold a semisquat posture during both a sudden loading task and a steady state load-support task. The two muscles studied were the multifidus and longissimus, and these two muscles were chosen because of the differences in their basic biomechanical functions (i.e., stabilizer vs. trunk extensor) (Bergmark, 1989; McGill, 1991; Wilke et al., 1995), leading to a potential differential response to the influence of the knee support device. From the more basic science perspective, the sudden loading condition was considered because it was reasoned that the "perturbing" nature of this type of loading (e.g., Lavender et al., 1989; Cresswell et al., 1994; Gagnon et al., 1995; Brown et al., 2003; Lawrence et al. 2005, 2006) would have a greater chance of eliciting a differential response as a function of the levels of the knee support variable.

As one considers a static, two-dimensional biomechanical model of a harvester in a semisquat/semistoop posture, it is not immediately evident that the insertion of a knee support would have any impact on the activation levels of the muscles of the low back—the device would not have an impact on the moment of the load, the angle of the torso, or the magnitude of the load, and so on. However, during a sudden loading event, the dynamic response of the system, as it attempts to reachieve equilibrium, might be affected by the loss of the degree of freedom of the ankle joint. A significant response to the knee support would indicate that the ankle joint plays an important role in the system response. This was the motivation for considering the sudden loading component of this study. Consistent with the previous studies noted above, the EMG response of the extensor musculature was evaluated in the period of peak muscle activation (0 to 1 s after sudden loading was experienced), but the results showed no significant effect of the knee support device during these sudden loading conditions.

From the applied perspective, the main objective of this study was to evaluate the effect of the knee support mechanism during the steady-state activity of a simple load-support task, as might be seen as a pepper harvester is gathering the produce. The load-relieving aspects of this device for the lower extremities are clear, but our a priori concern was that this intervention had the potential to negatively influence the muscle activation profiles of the trunk extensors, leading to concerns related to the implementation of such a device in the field. The biomechanical construct that led to this concern was centered on the concept that the knee support essentially eliminates one degree of freedom in the kinematic chain (the ankle joint), which could then focus more of the response on the remaining joints in the chain. Data from neither the steady-state phase nor the sudden loading phase of the exertions supported this loading model; therefore, it appears that these concerns were unwarranted. Follow-up research is underway to more completely explore these responses under realistic harvesting conditions.

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