Spinal loading during manual materials handling in a kneeling posture

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Abstract

Stooped, restricted, kneeling, and other awkward postures adopted during manual materials handling have frequently been associated with LBP onset. However, lift assessment tools have focused on materials handling performed in an upright, or nearly upright standing posture. Unfortunately, many of the tools designed to analyze standing postures are not easily adapted to jobs requiring restricted postures. Therefore, the objective of this study was to evaluate spinal loading during manual materials handing in kneeling postures and determine if those loads can be predicted using simple regression. An EMG-driven biomechanical model, previously validated for upright lifting, was adapted for use in kneeling tasks. Subjects knelt under a 1.07 m ceiling and lifted luggage of six weights (6.8, 10.9, 15.0, 19.1, 23.1 and, 27.2 kgf) to one of four destination heights (0, 25.4, 53.3, 78.7 cm). Spine loading was significantly affected by both destination height and load weight. Destination height increased compression, AP shear and lateral shear by an average of 14.5, 3.7 and 6.6 N respectively per cm height increase. Load weight increased compression, AP shear and lateral shear by an average of 83.8, 27.0 and 13.1 N respectively per kgf lifted. Regression equations were developed to predict peak spine loading using subject height, load weight and destination height with R^2 values of 0.62, 0.51 and 0.57 for compression, AP and lateral shear respectively.
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1. Introduction

Stooped, restricted, kneeling, and other awkward postures adopted during manual materials handling have frequently been associated with low back pain (LBP) onset [17,20,22,28,46]. Miners, aircraft baggage handlers, carpenters, mechanics and agricultural workers are among the many professions that require such postures [8,19].

To date however, the tools that have been developed for controlling LBP incidence in industry have focused on materials handling performed in an upright, or nearly upright standing posture [2,35,45,47]. Unfortunately, many of the tools designed to analyze standing postures are not easily adapted to jobs requiring restricted postures. Arguably the most well-known industrial assessment tool, the

1991 NIOSH lifting equation, specifically states: ''The revised lifting equation was not designed to assess tasks involving one-handed lifting, lifting while seated or kneeling, or lifting in a constrained or restricted work space'' [47]. Recommendations for work in restricted postures are therefore limited to trial-and-error learning on the job and general recommendations such as those made by Gallagher and Unger [12] who recommended a 14–18% decrease in weight of supply items when the kneeling posture must be used for lifting.

Lift assessment tools intended for general use must use data that is not difficult to collect but which is still associated with LBP incidence. The NIOSH lifting equation [44] and another well known industrial assessment tool developed by Marras and Granata [33] use (collectively) combinations of: lift origin height, lift destination height, load lifted, horizontal moment arm, load coupling, lift asymmetry, sagittal load moment, lifting frequency, trunk lateral

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velocity, trunk twisting velocity and trunk sagittal angle. Unfortunately, trunk kinematics require specialized instrumentation and the horizontal moment arm needed to determine load moment may be difficult to obtain in jobs requiring restricted or kneeling postures.

The reasons for the difficulty in adapting tools designed for upright postures to situations requiring restricted postures are, in part, related to changes in the soft tissue biomechanics and muscle strength. These posture-related changes may be due to changes in the muscle moment arm, alteration of muscle length, the reduced number of muscles in the hips and thighs contributing to the lifting moment, and/or decreased stability and mobility while kneeling [5,8,23].

Alterations in muscle recruitment patterns were reported by Gallagher et al. [10,11], who demonstrated that posture specific selective muscle recruitment occurs, independent of the load lifted, as individuals shift between standing, stooping, and kneeling on one or two knees. Squatting has been shown to reduce the activity of the lumbar erector spinae when compared to kneeling for a baggage handling task, while kneeling resulted in lower activity in the trapezius [46].

Several researchers have noted a reduction in strength between standing and kneeling on two knees [18,19] and between stooping and kneeling [8,10,12]. Where there is decreased strength, a person performing a given work task must work at a greater percentage of their maximal capacity, thus increasing the likelihood of an over-exertion type injury.

Given that forces from the active trunk muscles are the primary determinant of spinal load, increased muscle loading associated with restricted postures imposes increased spinal loading. Similarly, decreases in muscle moment arm accompanying flexion of the pelvis with respect to the lumbar spine will increase muscle force and thus, increase spinal loading [23]. In turn, increased spinal loading is associated with increased reporting of low-back disorders [3,41]. Lower ceiling height increased spinal compression in a simulation of baggage handling [43] and resulted in increased sagittal moment about the spine in a simulated mining task [9]. Gallagher et al. [9] found ceiling height may also have affected how subjects performed tasks, regardless of the posture assumed. Lower ceiling heights, 1.2 versus 1.5 m, caused a significant increase in spinal moment, irrespective of the posture used. These findings suggest that individual anthropometric differences may also play a role in determining spine loads where work postures are restricted by environmental constraints.

Muscle forces are frequently predicted from the electromyographic (EMG) activity levels recorded from the muscles, which in turn are used to quantify spine loads in EMG-assisted spine biomechanical models [35,39]. The adaptation of such models to kneeling postures warrants further study due to the changing biomechanical relationships such as decreases in muscle moment arms and changes in muscle length. Hence, in order to achieve the goal of this research, which was to evaluate spinal loading as individuals performed simulated work tasks in kneeling postures, the refinement of a previously developed EMG-assisted spine model [32] became a pre-requisite activity.

Given that the movement degrees of freedom and stability are limited when individuals are kneeling in a confined space, we theorized that the resulting spine loads may be determined largely by parameters such as load weight, load destination height and subject anthropometry. Thus, the purpose of this paper, in addition to describing the EMG-assisted model refinements necessary to accommodate kneeling postures, was to test the hypothesis that spinal loading is significantly effected by the weight handled and the height to which an object is lifted. Further, this paper investigated the use of these parameters in regression models that predict spinal loads. A regression model that could assist in evaluating lifting tasks while kneeling for low back disorder risk would be a valuable tool for those attempting to prevent low back disorders.

2. Methods

2.1. Subjects

Twelve healthy male college students with no prior history of low back disorders or knee problems volunteered to participate in the study. None of the participants had professional manual materials handling experience. The average (SD) age, stature and weight of the subject population was 24.6 (3.2) years, 180.8 (6.9) cm and 74.7 (10.0) kgf, respectively. The anthropometric data ranges of participants are presented in Table 1.

2.2. Task

Subjects performed a repetitive lifting task in a restricted posture while kneeling on a force plate. The restriction was imposed by a ceiling located 1.07 m above the floor which was based upon the height of the rear baggage compartment inside a Boeing 737 aircraft. Prior to this study, baggage handling in Boeing 737 aircraft was observed at an airport. Based on these observations, a piece of luggage was placed in a marked area located 45° clockwise from the subject with the midpoint of the luggage 45.7 cm from the force plate. The subject then moved the luggage to a

Table 1
Anthropometric data ranges of participants ($n = 12$)

Variable	Units	Mean	SD	Min	Max
Age	years	24.6	3.2	19	30
Stature	cm	180.8	6.9	170.5	195.4
Weight	kgf	74.7	10.0	54.4	93.0
Body mass index (BMI)	kgf/m^2	22.9	2.5	18.8	26.9
Trunk breadth at xyphoid	cm	30.0	2.5	26.4	35.1
Trunk depth at xyphoid	cm	21.5	2.4	17.4	25.1
Trunk breadth at pelvis	cm	27.6	1.8	24.4	31.4
Trunk depth at pelvis	cm	19.4	1.4	16.9	21.2

destination 45° counterclockwise and 63.5 cm from the force plate. The task, consistent with the baggage handling behaviors was repeated every six seconds. The subject and experimental setup are shown in Fig. 1. Subjects handled luggage of six different weights and placed them at four different heights. Each piece of luggage was a two-wheel suitcase measuring $45 \times 28 \times 66$ cm (L × W × H). Combinations of experimental conditions (6 weights, 4 heights, 2 repetitions) were presented to the subjects in a randomized order. After every five lifts the subject was allowed to stand and rest to reduce the potential for development of fatigue. Half way through the study subjects were given a longer break and allowed to move freely.

2.3. Independent variables

The two independent variables were *luggage destination height* (slide along floor from origin to destination, lift to 25.4, 53.3, 78.7 cm) and *luggage weight* (6.8, 10.9, 15.0, 19.1, 23.1, 27.2 kgf). Destination heights were chosen to represent a range from floor to shoulder height for a typical kneeling individual. The 27.2 kgf weight was chosen as the upper limit to provide a realistically heavy load that would not pose excessive risk to these subjects who were untrained in this type of task.

2.4. Dependent variables

Dependent measures in this experiment consisted of spinal loading, and trunk kinematics. The sampled muscles included the following bilateral muscle pairs: Latissimus Dorsi (LATR and LATL), Erector Spinae (ERSR and ERSL), Internal Oblique (IOBR and IOBL), External Oblique (EXOR and EXOL), and Rectus Abdominus (RABR and RABL).

A whole-body free-dynamic model that utilizes electromyographic (EMG) data as input was used to determine spine loading [31]. Spine loading variables consisted of maximum values of spine compression, spine anterior–posterior (AP) shear, and spine lateral shear. Maximum values were also determined for normalized muscle activity for each muscle. Trunk kinematic variables determined for each lift were maximum values for position, velocity, and acceleration in each plane (sagittal, lateral, and transverse). For the purpose of this experiment, a "lift" is defined as the time between when the subject first grasps the load and when the load is released.

2.5. Apparatus

2.5.1. EMG monitoring

Bipolar Ag–Cl surface electrodes were affixed to the skin over the muscles and connected to a lightweight preamplifier located near the electrodes. Electrode pairs had an interelectrode distance of 3 cm. The signal passed through shielded cables to a hardware rack where they were further amplified, high pass filtered at 30 Hz, low pass filtered at 1000 Hz and notch filtered at 60 Hz. The signals were then rectified and integrated via a 20 ms sliding window. Surface electrodes were placed over ERSR, ERSL, RABR, RABL,

Fig. 1. A subject in the apparatus used during experiment. From a designated position 45° to the subject's right, luggage was lifted and placed on the shelf to the subject's left.

LATR, LATL, EXOR, EXOL, and IOBR and IOBL in order to provide input to the EMG-assisted free-dynamic model. Specific electrode locations can be found in Fathallah et al. [7] and Mirka and Marras [40].

2.5.2. Lumbar motion monitor

A lumbar motion monitor (LMM) [30] was used to obtain the trunk kinematics. These data were used to track instantaneous trunk muscle length, as well as the geometric relationship between the thorax and the pelvis, throughout each exertion for use in the EMG-assisted model.

2.5.3. Force plate

Subjects knelt, unrestricted, on a force plate during the lifting tasks. To allow subjects to kneel and still be supported by the force plate, a 1.5 cm thick wooden plate was added, which extended beyond the back edge of the force plate. During kneeling exertions, load lifted was registered by the force plate and added to the spinal compression calculated by the EMG-assisted biomechanical model.

All data were monitored simultaneously using proprietary software, a National Instruments PCI-6031 E A/D board, and a 2 GHz Pentium IV computer. EMG signals, were sampled at 1000 Hz; all other signals were sampled at 100 Hz.

2.5.4. EMG-assisted biomechanical model

An EMG-assisted biomechanical model employed the EMG, kinetic, and kinematic data as inputs to compute dynamic loads on the spine [14,31,33,36,37]. EMG data were normalized relative to maximum contraction values. For normalization purposes, static maximum voluntary contraction (MVC) exertions were performed in each of six directions (trunk flexion, extension, right lateral, left lateral, right twist, and left twist) while standing in a structure that immobilizes the pelvis and the lower extremities [16]. MVCs were collected with subjects standing because peak trunk muscle activity has been shown to be equivalent in standing and kneeling postures [8].

Initially, standing and kneeling calibration exertions were then performed with the subject standing/kneeling on the force plate. Force plate data was used, along with a proprietary system, developed in-house [7], to determine moment of the torso relative to the L5/S1 joint during the calibration lifts and a subject-specific muscle gain was determined. A preliminary analysis indicated no difference between gains achieved using standing versus kneeling calibrations. However, there was concern that some subjects might naturally lean backwards on the wooden forceplate extension and create an inaccurate moment reading at the forceplate. To simplify data collection, standing calibrations only were used for the remaining subjects.

The model was then able to incorporate the normalized muscle activities, dynamic trunk motion, and subject muscle gain to determine the contractile forces of the 10 trunk muscles during the kneeling exertions. Spinal compression, lateral shear, and anterior–posterior (AP) shear forces were computed from the vector sum of the muscle forces with the addition of load weight to compression. Thus, three-dimensional dynamic spinal loads were determined for each lifting exertion. The data collection methods, biomechanical model structure, and validation for standing lifting activity have been published previously [13,16,31,33,36–38].

In order to adapt the model for the restricted kneeling posture used in this research, a number of modifications were applied. During some experimental conditions muscle activations exceeded the levels observed during the isometric maximal exertions described previously. When this occurred, the subject's MVC values were replaced with maximum values observed during the experimental conditions for normalization purposes. In addition, model muscle gain (muscle force producing capability per cubic centimeter) was calculated using the average of four sagittal lifts performed while the subject stood unrestricted on the force plate. In order to keep the muscle force contribution within the active portion of the length-strength relationship, muscle length was not allowed to exceed 160% of resting muscle length [6,24–26,48].

2.6. Statistical analysis

Statistical analysis was conducted using SAS (SAS Institute, Cary, NC) for analysis-of-variance (ANOVA) tests for significant differences between means for luggage weight and destination height. Diagnostics were performed on all data in order to ensure statistical model assumptions were not violated (error mean of zero, constant variance, normal distribution and independence). In the case of spinal loading the ratio of the largest sample variance to the smallest sample variance was well above 3, the rule of thumb for violation of the constant variance assumption. Natural log transformations were then performed in order to restore model validity. Once significance was determined, the data were transformed back for more intuitive presentation here.

Regression analyses were performed using stepwise regression procedure to predict the directional spinal loadings as a function of luggage weight and destination height. The probability for terms to enter or leave the regression model was set at $p = 0.25$.

3. Results

3.1. Spinal loading

The main effects of luggage weight and lift destination height on peak compression, AP shear and lateral shear forces were significant at a $p < 0.0001$ level. As shown in Fig. 2, peak compression at L5/S1 increased from 1623.8 N for the lowest luggage weight to 3334.1 N for the heaviest luggage weight. Peak AP shear force increased from 357.5 N to 909.2 N. Peak lateral shear force also increased from 178.2 N to 446.0 N. Post hoc tests showed that spinal loading at each luggage weight level was significantly different
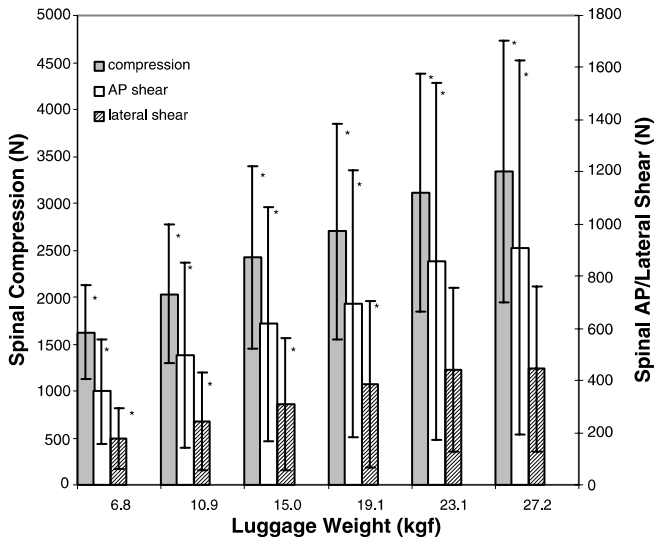
Fig. 2. The effect of luggage weight on peak compression, AP shear and lateral shear forces (mean and SD). '*' indicates significantly different spinal loading from other weight levels ($p < 0.05$).

from that at other weight levels ($p < 0.05$), except that lateral shear forces at luggage weights of 23.1 and 27.2 kgf were not statistically different from each other ($p = 0.355$).

Spinal loading also increased as lift destination height increased (Fig. 3). Peak compression, AP shear, and lateral shear were 1463.9 N, 397.0 N, and 225.3 N, respectively, for the floor level slide. These values increased to 3248.2 N, 842.1 N, and 594.9 N for the highest level of lift. The compression force at each lift destination height was significantly different from that at other heights ($p < 0.0001$). The AP shear forces at floor level (0 cm) and the highest level (78.7 cm) were significantly different from those at other lift destination heights ($p < 0.05$). But AP shear forces at 25.4 and 53.3 cm lifting heights were
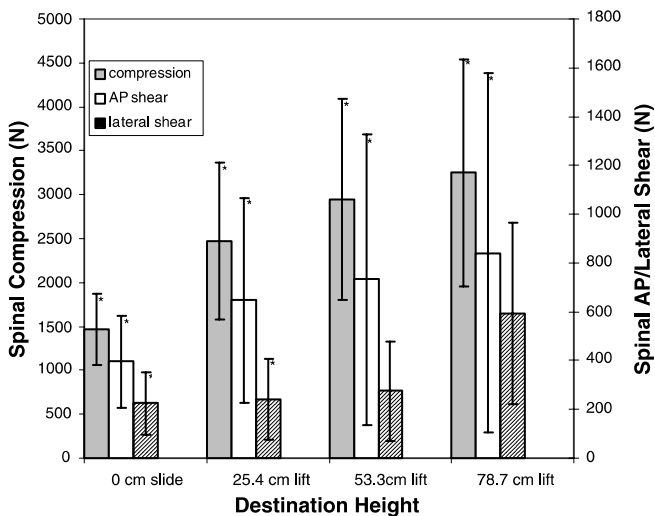
not statistically different from each other ($p = 0.136$). For lateral shear force, the floor level slide and 25.4 cm lift were not different from each other ($p = 0.927$). However, the forces at the higher levels of lift destination height were significantly different from those of other lift heights ($p < 0.05$).

For each weight level, peak compression increased with increasing lift destination height. Likewise, for each lift destination height, peak compression increased with increasing luggage weight. As shown in Fig. 4, for the floor level slide (0 cm), the difference was less than 630 N between the 6.8 kgf luggage and the 27.2 kgf luggage. However, as there was no 0 cm lift exertion for comparison to the slide data, further comparisons cannot be made.

3.2. Kinematics

Increasing the luggage weight reduced average amount of forward trunk flexion, the average twisting velocity and acceleration. At the same time increasing the luggage weight increased both the sagittal plane velocity and the maximum sagittal acceleration. Likewise significant increases were observed in the maximum and average maximum and average lateral flexion angle, velocity, and accelerations. Between the lowest and highest weights, the maximum sagittal accelerations increased from 66.1 deg/s^2 to 86.7 deg/s^2 (a 30% increase) while the maximum lateral acceleration increased from 90.5 deg/s^2 to 111.0 deg/s^2 (a 23% increase).

As the lift destination height increased, the maximum and average sagittal angle decreased. Increasing the lift destination height increased the maximum and average lateral angle, twisting angle, sagittal velocity, average lateral velocity, and sagittal acceleration. In addition and maximum lateral acceleration increased. The most significant change was in maximum sagittal acceleration, which increased from 53.7 deg/s^2 for the lowest height to 94.6 deg/s^2 for the highest height, a 76% change. Maximum sagittal velocity also

Fig. 3. The effect of lift destination height on peak compression, AP shear and lateral shear forces (mean and SD). '*' indicates significantly different spinal loading from other heights ($p < 0.05$).
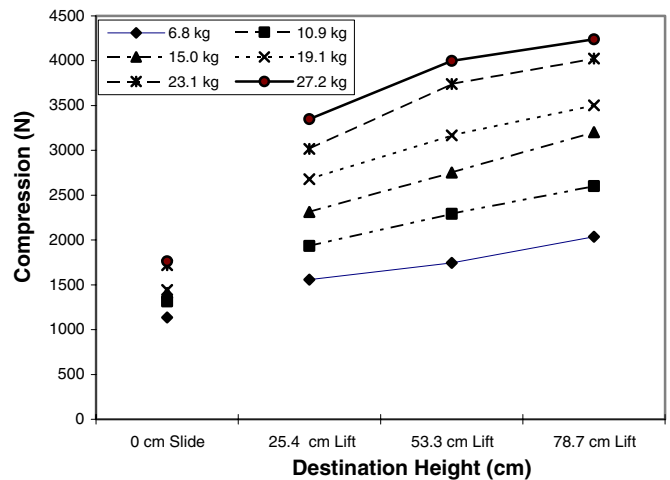

Fig. 4. The effect of luggage weight and lift destination height interaction on peak compression (mean values).

increased by 91% (from 16.5 deg/s to 31.5 deg/s) as the lift destination height changed from the lowest to the highest.

The luggage weight and lift destination height interaction was significant for maximum sagittal acceleration ($p < 0.01$). This effect was probably due to the significant increase in maximum sagittal acceleration when subjects lifted the two heaviest weights to the highest height.

4. Regression analysis

The following regression equations for peak compression, AP, and lateral shear force were obtained using the subjects' standing height, luggage weight, and lift destination height:

$$\text{Compression} = e^{5.282 + 0.008*\text{standing height} + 0.033*\text{luggage weight} + 0.009*\text{lift destination height}} \tag{1}$$

$$\text{APshear} = e^{-3.456 + 0.048*\text{standing height} + 0.042*\text{luggage weight} + 0.007*\text{lift destination height}} \tag{2}$$

$$\text{Lateralshear} = e^{2.493 + 0.010*\text{standingheight} + 0.045*\text{luggageweight} + 0.011*\text{liftdestinationheight}} \tag{3}$$

where standing height is the subject's stature when standing (cm); luggage weight is the weight of the luggage lifted (kgf) and lift destination height is the vertical height of the lift measured from the floor to the destination (cm).

All the equations were significant at $p < 0.0001$. The R^2 value was 0.62 for the compression equation. However, the R^2 for AP shear force and lateral shear were only 0.46 and 0.38, respectively.

In an attempt to improve the R^2 value for predicting AP and lateral shear, relevant kinematic variables were included in the regression model and the following equations were obtained:

$$\text{AP shear} = e^{-1.477 + 0.029*\text{standing height} + 0.043*\text{luggage weight} + 0.007*\text{lift destination height} + 0.024*\text{MSF}} \tag{4}$$

$$\text{Lateral shear} = e^{2.363 + 0.009*\text{standing height} + 0.037*\text{luggage weight} + 0.005*\text{lift destination height} + 0.055*\text{MLF}} \tag{5}$$

where MSF is the maximum sagittal flexion during the lift task in deg and MLF is the maximum lateral flexion during the lift task in degrees.

The R^2 values for predicting AP and lateral shear forces increased to 0.51 and 0.57, respectively.

5. Discussion

Of primary importance to this research is improving the understanding of the biomechanical loads placed on the spine during the performance of material handling tasks in awkward postures such as kneeling. Specifically, this study has attempted to further our understanding of how two common workplace parameters, luggage weight and lift destination height, affect the biomechanical loading of the spine while in kneeling postures. In nearly all instances, spinal loading increased with luggage weight and lift destination height. This includes spinal compression, AP shear

and lateral shear forces. These results were largely due to the increases in muscle activity that were generally observed with increasing luggage weight and lift destination height.

There are several variables that could impact the generalizability of these results. First, the environmental variables constraining the work posture must be considered. For example, in this study the work area was vertically constrained. Gallagher et al. [9] showed that the peak lumbar moment was dependent on the ceiling height in their study of cable lifting. Had this constraint not been in place the taller individuals would potentially have shown less forward spine flexion at the completion of the lift. Thus, changing this constraint would be expected to change the variance accounted for by anthropometric variables in the regression models. Second, the physical and behavioral characteristics of the lift, for example the lift origin and destination heights, the horizontal distance, the object weight, asymmetry, and the number of hands used have been shown to impact the biomechanical loadings in standing postures [1,21,27,29,42,44].

In addition, the subject population used here consisted of college-age male students who were inexperienced at manual materials handling tasks. In contrast, the population of actual industrial workers who perform lifting tasks is comprised of both genders of varying ages, accumulating years of handling experience. Granata and colleagues [15] found increased spinal compression, AP and lateral shear in experienced lifters relative to inexperienced lifters performing the same task. Therefore, the spinal loading characteristics may become more hazardous in industrial situations where work is performed by experienced individuals. How these factors impact spinal loads when working in a kneeling posture is not well understood.

This research raises the further question of how tasks performed standing differ in terms of spinal loading from kneeling tasks. Our laboratory has used the adapted model to compare kneeling lifts to similar lifting performed while standing [49]. The results of the aforementioned study indicated significant increases in lateral and AP shear forces when kneeling.

Previous findings with the EMG-driven model for standing subjects have shown a monotonically increasing relationship between load weight and spinal compression

in for stationary subjects [34]. In that research the average of the peak spine compression value for male subjects when lifting 27.3 kgf from a height of 95 cm (hip level) was approximately 2500 N. In the current study employing a kneeling posture, the average of the peak spine compression value across the three actual lifting conditions with the 27.2 kgf load was 3861 N. In other words, lifting a nearly equivalent weight from approximately the same level relative to L5/S1 yielded a spine compression value that was nearly 50% greater when lifting from a kneeling posture as compared with a standing posture.

When the subject is allowed to move their feet, spinal compression did not increase for 3–9 kg load increases at lower weights [4]. This research indicated that trunk dynamics could be used as a buffer for small increases in load weight. Whether this relationship would be true for kneeling subjects is unknown. Kneeling is an inherently more limited posture than standing. Subjects are not able to move with the same dexterity as when standing and may not be able to use their trunk dynamics in the same way.

Also of interest however, is how vertical destination height impacted spinal loads. Independent of how much weight was lifted in the non-slide conditions, spinal compression increased with destination height. Shear forces also tended to increase. Similarly, the kinematic data were affected by both the luggage weight and lift destination height. Of note is the increase in maximum sagittal and lateral accelerations observed with increasing lift destination height. This suggests a ballistic lifting motion wherein subjects tried to increase the momentum of the luggage at the beginning of the lift in order to reduce the load during the latter part of the lift, thus reducing the strength demands on the shoulders as the shoulder flexion angle, and the load moment increased.

5.1. Data collection and modeling challenges

Sampling and modeling EMG activity of the trunk muscles in a kneeling posture created a new set of challenges during this investigation. The model that was used to quantify spinal loading and kinematics is largely driven by the muscular activity obtained through EMG. This model, developed and validated through lifting studies in an upright posture [14], relies on electrical activity being compared to a set of MVCs. The kneeling tasks performed in this study resulted in samples of muscle activity that were greater than the activity obtained from the MVCs. When this occurred the maximal values obtained during the lifting task were used in place of the "maximum" EMG values obtained from the initial set of isometric exertions. This approach resulted in reasonable muscle gain values (mean: 41.9, SD: 15.6) and high (0.90 or greater) R^2 values when comparing calculated internal and external sagittal plane moment during the calibration exertions.

The cause of the EMG values greater than MVC appears to stem from the posture used in this study. Sim-

ilar standing dynamic exertions do not often result in EMG values exceeding MVC using healthy subjects. Kneeling appeared to create a flexion in the pelvis relative to the lumbar spine that would increase muscle length and possibly drive the muscle into an unfavorable length-strength relationship. The height restriction may have also caused subjects to accelerate the load more to reach its destination, resulting in increased muscle velocity.

As expressed in the methods, a number of changes to the standing model were necessary in order to accommodate the kneeling posture. Consistent with the literature muscle length was not allowed beyond 160% of resting length [6,24–26,48]. The dynamic model used to calculate spinal loads takes into account the ability of muscle to produce force at various lengths due to both active and passive muscle characteristics. Originally, the values used to determine this ratio in the model increased indefinitely with an increasing muscle length. It was deemed necessary to limit the length modulation factor at 1.9, or 160% (as seen in Fig. 5) of muscle resting length after instrumentation recorded muscle lengths greater than this, resulting in unusually high spinal load forces due to a steep increase in the modulation curve.

Muscular gain values used in model calculations were set at a fixed value, derived during a set of standardized kneeling mid-sagittal plane lifts. Gain is defined as muscle force per unit area with accepted physiological values range from 30 to 100 N/cm^2 [32]. Standing and kneeling calibration exertions were analyzed for each subject. Calculated muscle gains were similar for standing and kneeling with magnitudes within the physiological range.

Thus, there are many physical variables that arise when trying to model humans in restricted postures. These include spatial dimensions, lift origin and destination and individual subject parameters such as height and physical strength. The data presented are intended to simulate only one particular task. Thus, the model was adapted using as much of a scientific foundation as feasible to asses kneeling tasks, however future studies are needed to explore the effects of various restricted postures on the human body.
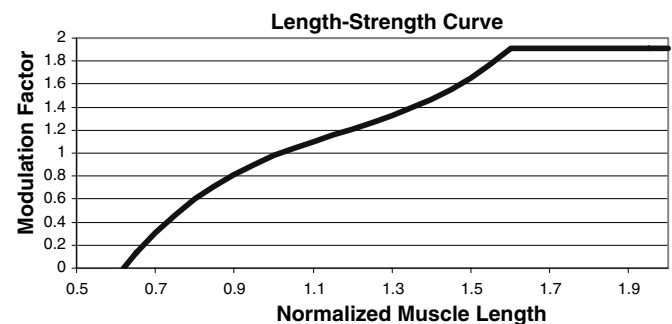

Fig. 5. Length–strength modulation function used in the EMG driven biomechanical model.

5.2. Estimating effects by regression

Our hypothesis was that an empirical model could be developed to predict spine loads based on readily observable parameters describing lifts from kneeling postures. It should be recognized that the models developed were based on a single constrained vertical height condition. Using the data from 12 subjects, regression models were developed to estimate spinal compression, anterior–posterior shear and lateral shear forces on the L5/S1 joint when lifting in a constrained kneeling posture. The goal of this analysis was not to provide a definitive workplace analysis tool but rather to explore the potential for such a model.

A number of possible regressors were evaluated using a stepwise procedure. Lift height, luggage weight and subject height proved to be the most significant combination of variables for each analysis, which also happen to be the easiest to obtain, requiring no special measurement devices beyond a scale and tape measure. Other variables tested included trunk angles (i.e. sagittal bending, lateral bending and twisting), trunk velocities and accelerations. While shear estimates improved with the inclusion of trunk kinematics, angular data is difficult to accurately estimate visually.

The variables used in the regression equations were all chosen using stepwise regression without considering their biomechanical appropriateness. Therefore, we considered the relationship between the variables chosen using the stepwise technique and their effect on spinal loads. Lift destination height can influence spine posture and the corresponding muscle length-strength relationship, muscle activation patterns, and load acceleration characteristics. In this study, lift destination height shows a positive linear association with variables such as compression ($r = 0.54$) and lateral shear ($r = 0.45$). Object weight lifted in a constrained posture has a direct effect on muscle activity levels and is positively correlated with compression, AP shear and lateral shear ($r = 0.49, 0.35, 0.35$). Subject height positively correlates with maximum lateral position ($r = 0.62$). Taller subjects were constrained more by the height restriction and forced to bend forward and to the side more in order to complete the lifts. In addition, the larger torso length of taller subjects would increase spinal loading more as the subjects flexed.

It should be recognized that changes in the height restriction of the workspace away from the 107 cm value used here may change the contribution of subject height. One could estimate that an increase in ceiling height may decrease the overall spinal loading due to a more upright posture by reducing the moment created by the bent spine throughout the lift. On the other hand, maximum spinal flexion typically occurred at the initiation of the lift and was not affected by the vertical constraint. Therefore a change in height restriction may not affect the peak measures of spinal loading.

A second set of regression models of AP and lateral shear included measured postural variables in an attempt to improve the reliability of predicting spinal loads. The addition of these variables resulted in improved model performance. This indicates that more accurate predictions may be possible if the capability of collecting complex kinematic data exists.

6. Conclusions

The spine loads in kneeling postures are independently sensitive to both the weight lifted and the final destination height. Increasing weight increased spinal compression, AP and lateral shear by 83.8, 27.0 and 13.1 N/kgf respectively. Similarly, increasing destination height increased spinal compression, AP and lateral shear by 14.5, 3.7 and 6.6 N/cm respectively. These effects can be used to estimate spinal loading using easy to measure variables such as subject height, lift destination height and load weight.

References

[1] Allread WG, Marras WS, Parnianpour M. Trunk kinematics of one-handed lifting, and the effects of asymmetry and load weight. Ergonomics 1996;39(2):322.

[2] Chaffin DB, Baker WH. A biomechanical model for analysis of symmetric sagittal plane lifting. AIIE Trans 1970;2(1):16–27.

[3] Chaffin DB, Park KS. A longitudinal study of low-back pain as associated with occupational weight lifting factors. Am Ind Hyg Assoc J 1973;34(12):513–25.

[4] Davis KG, Marras WS. Assessment of the relationship between box weight and trunk kinematics: does a reduction in box weight necessarily correspond to a decrease in spinal loading?. Hum Factors 2000;42(2):195–208.

[5] Drury CG. Influence of restricted space on manual materials handling. Ergonomics 1985;28(1):167–75.

[6] Elftman H. Biomechanics of muscle with particular application to studies of gait. J Bone Joint Surg Am 1966;48(2):363–77.

[7] Fathallah FA, Marras WS, Parnianpour M. The effect of complex dynamic lifting and lowering characteristics on trunk muscles recruitment. J Occupat Rehab 1997;7:121–38.

[8] Gallagher S. Trunk extension strength and muscle activity in standing and kneeling postures. Spine 1997;22(16):1864–72.

[9] Gallagher S, Hamrick CA, Cornelius K, Redfern MS. The effects of restricted workspace on lumbar spine loading. Occup Ergon 2001;2(4):201.

[10] Gallagher S, Marras WS, Bobick TG. Lifting in stooped and kneeling postures: Effects on lifting capacity, metabolic costs, and electromyography at eight trunk muscles. Int J Ind Ergon 1988;3(1):65–76.

[11] Gallagher S, Marras WS, Davis KG, Kovacs K. Effects of posture on dynamic back loading during a cable lifting task. Ergonomics 2002;45(5):380–98.

[12] Gallagher S, Unger RL. Lifting in four restricted lifting conditions. Appl Ergon 1990;21(3):237–45.

[13] Granata KP, Marras WS. An EMG-assisted model of trunk loading during free-dynamic lifting. J Biomech 1995;28(11):1309–17.

[14] Granata KP, Marras WS, Davis KG. Biomechanical assessment of lifting dynamics, muscle activity and spinal loads while using three different styles of lifting belt. Clin Biomech (Bristol, Avon) 1997;12(2):107–15.

[15] Granata KP, Marras WS, Davis KG. Variation in spinal load and trunk dynamics during repeated lifting exertions. Clin Biomech (Bristol, Avon) 1999;14(6):367–75.

[16] Granata KP, Marras WS, Fathallah FA. A method for measuring external loads during dynamic lifting exertions. J Biomech 1996;29(9):1219–22.

[17] Harkness EF, Macfarlane GJ, Nahit ES, Silman AJ, McBeth J. Risk factors for new-onset low back pain amongst cohorts of newly employed workers. Rheumatology (Oxford) 2003;42(8):959–68.

[18] Haslegrave CM, Tracy MF, Corlett EN. Force exertion in awkward working postures–strength capability while twisting or working overhead. Ergonomics 1997;40(12):1335–62.

[19] Haslegrave CM, Tracy MF, Corlett EN. Strength capability while kneeling. Ergonomics 1997;40(12):1363–79.

[20] Holmstrom EB, Lindell J, Moritz U. Low back and neck/shoulder pain in construction workers: occupational workload and psychosocial risk factors. Part 1: Relationship to low back pain. Spine 1992;17(6):663–71.

[21] Jager M, Luttmann A. The load on the lumbar spine during asymmetrical bi-manual materials handling. Ergonomics 1991; 35(7-8):783–805.

[22] Jorgensen K, Jensen B, Stokholm J. Postural strain and discomfort during loading and unloading flights. Trends Ergon/Hum Factors IV 1987:663–73.

[23] Jorgensen MJ, Marras WS, Gupta P, Waters TR. Effect of torso flexion on the lumbar torso extensor muscle sagittal plane moment arms. Spine J 2003;3(5):363–9.

[24] Kroemer KH. Human strength evaluation. In: Karwowski W, Marras WS, editors. The occupational ergonomics handbook. CRC Press; 1999.

[25] Kroemer KH, Kroemer HJ, Kroemer-Elbert KE. Engineering physiology. Bases of human factors/ergonomics. Third ed. New York: Van Nostrand Reinhold; 1997.

[26] Kroemer KHE, Marras WS, McGlothlin JD, McIntyre DR, Nordin M. On the measurement of human strength. Int J Ind Ergon 1990;6(3):199–210.

[27] Lavender S, Andersson GB, Schipplein OD, Fuentes CS. The effects of initial lifting height, load magnitude, and lifting speed on the peak dynamic L5/S1 moments. Int J Ind Ergon 2003;31(1):51–9.

[28] Lawrence JS. Rheumatism in coal miners. III. Occupational factors. Br J Ind Med 1955;12(3):249–61.

[29] Marras WS, Davis KG. Spine loading during asymmetric lifting using one versus two hands. Ergonomics 1998;41(6):817.

[30] Marras WS, Fathallah FA, Miller RJ, Davis SW, Mirka GA. Accuracy of a three-dimensional lumbar motion monitor for recording dynamic trunk motion characteristics. Int J Ind Ergon 1992;9:75–87.

[31] Marras WS, Granata KP. A biomechanical assessment and model of axial twisting in the thoracolumbar spine. Spine 1995;20(13):1440–51.

[32] Marras WS, Granata KP. The development of an EMG-assisted model to assess spine loading during whole-body free-dynamic lifting. J Electromyogr Kinesiol 1997;7(4):259.

[33] Marras WS, Granata KP. Spine loading during trunk lateral bending motions. J Biomech 1997;30(7):697–703.

[34] Marras WS, Granata KP, Davis KG, Allread WG, Jorgensen MJ. The effects of box features on spinal loading during warehouse order selecting. Ergonomics 1999.

[35] Marras WS, Lavender SA, Leurgans SE, Rajulu SL, Allread WG, Fathallah FA, Ferguson SA. The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders. The effects of workplace factors, trunk position, and trunk motion characteristics on risk of injury. Spine 1993;18(5):617–28.

[36] Marras WS, Mirka GA. Muscle activities during asymmetric trunk angular accelerations. J Orthop Res 1990;8(6):824–32.

[37] Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: I. Model structure. Hum Factors 1991;33(2):123–37.

[38] Marras WS, Sommerich CM. A three-dimensional motion model of loads on the lumbar spine: II. Model validation. Hum Factors 1991;33(2):139–49.

[39] McGill SM, Norman RW. Partitioning of the L4–L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine 1986;11(7):666–78.

[40] Mirka GA, Marras WS. A stochastic model of trunk muscle coactivation during trunk bending. Spine 1993;18(11):1396–409.

[41] Norman R, Wells R, Neumann P, Frank J, Shannon H, Kerr M. Ontario Universities Back Pain Study G. A comparison of peak vs cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry. Clin Biomech 1998:561.

[42] Plamondon A, Gagnon M, Gravel D. Moments at the L (5)/S (1) joint during asymmetrical lifting: effects of different load trajectories and initial load positions. Clin Biomech (Bristol, Avon) 1995;10(3):128–36.

[43] Ruckert A, Rohmert W, Pressel G. Ergonomic research study on aircraft luggage handling. Ergonomics 1992;35(9):997–1012.

[44] Schipplein OD, Reinsel TE, Andersson GB, Lavender SA. The influence of initial horizontal weight placement on the loads at the lumbar spine while lifting. Spine 1995;20(17):1895–8.

[45] Snook SH, Ciriello VM. The design of manual handling tasks: revised tables of maximum acceptable weights and forces. Ergonomics 1991;34(9):1197–213.

[46] Stålhammar HR, Leskinen TPJ, Kuorinka IAA, Gautreau MHJ, Troup JDG. Postural, epidemiological and biomechanical analysis of luggage handling in an aircraft luggage compartment. Appl Ergon 1986;17(3):177–83.

[47] Waters TR, Putz-Anderson V, Garg A, Fine LJ. Revised NIOSH equation for the design and evaluation of manual lifting tasks. Ergonomics 1993;36(7):749–76.

[48] Watkins J, Structure and function of the musculoskeletal system, 1999, Champaign, IL.

[49] Yang G, Splittstoesser RE, Knapik GG, Trippany DR, Vatan Korkmaz S, Hoyle JA, Lahoti P, Lavender SA, Sommerich CM, Marras WS. Comparison of spinal loads in kneeling and standing postures during manual materials handling. Proc Hum Factors Ergon Soc 2005;49.

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