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Comparison of the 1991 NIOSH Lifting Equation and Erector Spinae Muscle Electromyography

by

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University of Ottawa

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Dedication

To my parents, Bruce and Irene, in appreciation of their unending support.

Abstract

This study determined whether lifts rated as acceptable by the 1991 NIOSH equation calculations elicited myoelectric amplitudes of the erector spinae musculature (ES) within acceptable muscular load limits for continuous repetitive lifting tasks. Ten male subjects had surface electrodes placed bilaterally along the spine at levels T9, L1 and L3, 4 cm, 9 cm and 3 cm lateral to the midline, respectively. Each subject performed eight trials of five lifting conditions that were used to examine the horizontal factor (HF) and asymmetrical factor (AF) of the 1991 NIOSH equation. All lifts were ordered randomly and initiated and terminated in a standing position. The lifting motion was unconstrained and incorporated a "flatback", free-style lifting technique. EMG data were collected and linear envelopes (LE) were ensemble averaged across subject trials for each condition and normalized to a maximum voluntary contraction (MVC). Subject LE EMGs were ensemble averaged to generate condition LE EMG averages and subsequently converted to amplitude probability distribution functions (APDF). Percent MVC values from the APDF curves were compared to muscular load limits. A three-way, repeated measures, mixed model analysis of variance determined significant main effects for conditions, electrode placements and probability levels of the APDFs. There was general agreement between the 1991 NIOSH equation and muscular load limits. Bilateral T9 ES often exceeded "static" muscular loads and right L3 often exceeded "static" and "median" muscular loads. There was a significant ($p < 0.01$) difference for each main effect. The APDF EMG analysis was more sensitive to differentiating between conditions than the 1991 NIOSH equation. Phasic and amplitude

EMG analysis of the ES for occupational lifting tasks could be best represented by the musculature at L3, 3 cm lateral to the midline.

Acknowledgments

The author would like to thank Dr. Pete Stothart and Dr. Gord Robertson for their time and energy towards co-supervising this study. Thanks also go to Dr. Marc Gessaroli for his consultation on the statistical design and analysis, and Gord Evans for providing some technical support. The author is grateful to Grass Instruments for the bioamplifiers that were supplied to collect data. Much appreciation goes to Corrie Powell for her support and help to organize the data. Finally, great thanks go to the subjects who freely gave their time and volunteered to be tested.

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1.0 Introduction

For decades researchers have made substantial efforts towards establishing standards for lifting tasks which will reduce the chances of injury, particularly to the lumbar spine region. Workers engaged in manual materials handling, are predisposed to the threat of low back pain (Hettinger, 1985). Chaffin and Andersson (1991) have stated that greater workloads increase mechanical stress to the musculoskeletal system and are therefore, an important factor in the cause of low back pain. Attempts to quantify safer working environments have led to the development of lifting standards by several occupational health research bodies. The National Institute for Occupational Safety and Health (NIOSH) has been a long-standing leader in the field of occupational lifting research and, in 1981, produced a lifting standard, designed to help protect the worker. The standard, in the form of an equation, satisfied sagittal, two handed lifting conditions and was based on biomechanical, physiological, and psychophysical considerations (Ayoub *et al.*, 1982).

Since then, research findings and practical experience have led to a revised 1991 NIOSH lifting standard equation which is compared to the 1981 equation in Table 1. In response to the preference of engineering methods over administrative controls, the 1991 equation contains, in addition to the 1981 equation factors, provisions for asymmetrical lifting, differences in hand-container couplings, as well as, more detailed work durations and lifting frequencies (Waters *et al.* 1993). Some major discrepancies between the 1981

NIOSH Guidelines and the 1991 NIOSH equation modifications include the use of a standard lifting location (SLL), reduction of the load constant from 40 kg to 23 kg, and the concept of a lifting index (LI) rather than the threshold limit values which classified three levels of control for any given lift (Putz-Anderson and Waters, 1991). Consistent with the 1981 NIOSH equation, the 1991 equation operates on the notion that the risk of lifting-related low back pain increases as the demands of the lifting task increase (Waters *et al.*, 1993).

TABLE 1. Comparison of equation factors for the 1981 and 1991 NIOSH equations (from Putz-Anderson and Waters, 1991)

Lifting Factors	1981 Equation	1991 Equation
Load constant	40 kg	23 kg
Horizontal (HF)	15/H	25/H
Vertical (VF)	1-.004[V-75]	1-.003[V-75]
Distance (DF)	0.7+7.5/D	0.82+4.5/D
Frequency (FF)	$1\text{-}F/F_{max}$	from Table
Asymmetry (AF)	not available	1-.0032A
Coupling (CF)	not available	from Table

The 1991 equation itself is comprised of six multipliers: horizontal, vertical, distance, asymmetric, coupling, and frequency. Kumar and Garand (1992) have found the horizontal and asymmetric lifting factors to cause the greatest increases in lumbosacral stress for infrequent lifting. Although the 1991 NIOSH equation is constructed from physiological, psychophysical and biomechanical knowledge, lumbosacral stress is a biomechanical measure in the absence of biochemically related tissue fatigue. Thus, the biomechanics criterion measure of 3400 N compression at the L5/S1 disc is the prominent criterion for the horizontal and asymmetric multipliers during low frequency lifting (Ayoub, 1991; Waters *et al.*, 1993). Development of the horizontal factor equation was dependant upon a 2D static lifting model (Chaffin, 1969). This model used a single equivalent muscle, located 5 cm posterior to the L5/S1 disk centre, to represent the extensor musculature of the back. Also included in the model is the intraabdominal pressure, which remains a contentious issue due to its limited trunk extensor force contribution (McGill and Norman, 1986; Marras and Mirka, 1990). The asymmetrical factor was determined from limited psychophysical data and supporting biomechanical data which together report a 30% reduction in maximal lifting capacity for asymmetric lifting tasks of 90° (Bean *et al.*, 1988).

Industrial lifting activities are, by their very nature, repetitious and controlled. One can reason that knowledge of the lifting factors' effects on the musculoskeletal system will lead to practical implementations of safety precautions, such as, load restrictions and adherence to safe posture. Major work on the affected musculoskeletal tissues during various lifting scenarios has revealed greater understanding of the interactions between different tissue structures and their contributions to the lifting effort (Seroussi and Pope, 1987; Marras and Mirka, 1988, 1989, 1990; McGill, 1991a, 1991b; Kumar and Garand, 1992; Lavender *et al.*, 1992; Sommerich and Marras, 1992). The general consensus among researchers is that within the boundaries of industrial lifting tasks, the lumbar muscle tissues make up the majority of the lifting moment. For many types of occupational lifts, if the spine's natural lordotic curve is preserved during the lift, the extensor force value may be composed almost entirely of muscular effort as the spinal ligaments remain disengaged

(McGill and Norman, 1986; Rosenburg and Seidel, 1989; Potvin *et al.*, 1991). Levels of lumbar extensor muscle activity can therefore be associated to compressive stress in the lumbar spine (Andersson *et al.*, 1977; Freivalds *et al.*, 1984; McGill and Norman, 1986; Seroussi and Pope, 1987).

Muscular load can be identified through measurement of the myoelectric signal (EMG) recorded with surface electrodes. Amplitudes of the EMG can indicate the level of contraction of the muscle being recorded (Jonsson, 1978). The use of EMG to quantify muscular load levels has been utilized through the amplitude probability distribution function (APDF) for ergonomic analysis of worksite tasks (Hagberg and Jonsson, 1975; Jonsson, 1982; Andersson and Ortengren, 1984; Winkle and Bendix, 1986; Louhevaara *et al*, 1990). The APDF defines the amplitudes of EMG values as a percent of the total number of occurrences of EMG values (Jonsson, 1976, 1978). An EMG recording normalized to a maximum voluntary contraction (MVC) can be plotted as a probability, from 0 to 1, describing the occurrence of each percent MVC (% MVC). The threshold limit APDF is defined by three levels of muscular contraction. The static level of muscular contraction is defined by a probability of 0.1 and must be no more than 2-5% MVC, The median level is defined by a probability of 0.5 and must be no more than 10-14% MVC. The peak level is defined by a probability of 0.9 and must be no more than 50-70% MVC. Based on assumptions of muscular endurance, exceeding these probability levels is unacceptable and suggests that the task under investigation be modified to reduce the muscular loads (Jonsson, 1978, 1982).

To date, there are no studies that quantify the safe level of lumbar muscle activation during standardized dynamic lifting tasks. Buckle *et al.* (1992) have stated that a single manual materials handling guideline is rarely adequate to fully assess a particular lifing task. According to Putz-Anderson and Waters, (1991) and Waters *et al.* (1993), validation of the 1991 NIOSH equation remains a priority. In biomechanics direct validation is not feasible in most cases. A popular validation technique is to compare the results of two different methodologies, both designed to measure the same parameter, and determine whether they arrive at the same conclusions.

The purpose of this study was to determine whether there was agreement between acceptable lifts determined from the 1991 NIOSH equation and muscular load limits of the m. erector spinae (ES). It was also possible to evaluate the effectiveness of the 1991 NIOSH equation to adjust the load for different lifting scenarios based on the amplitude response of the ES myoelectric activity.

2.0 Methodology

2.1 Introduction

The ES EMGs of 10 male volunteers were analysed during acceptable lifting conditions that were based on the NIOSH lifting equation. The raw EMG signals from eight lifting trials for each of five lift conditions were converted to ensemble averaged linear envelopes (LE) for each subject under each lifting condition and normalized to their MVC. Condition mean APDFs were created using subject averages of the LE EMG. The APDFs were delimited to the 1991 NIOSH horizontal factor (HF) and asymmetrical factor (AF) variables. The condition mean static, median, and peak %MVC values, defined by the APDF probability levels of 0.1, 0.5 and 0.9, respectively, were compared with Jonsson's (1978, 1982) muscular load limits. A mixed-model, repeated measures, 3-way analysis of variance (ANOVA) was performed to determine subject, lift condition, APDF probability and electrode placement effects. A comprehensive study of the results compared Jonsson's (1978, 1982) muscular load limits to the 1991 NIOSH equation for occupationally acceptable lifts.

2.2 Subjects

Ten male volunteers participated in the study after signing consent forms. All subjects were screened for participation in this study by completing the Par Q questionnaire and reporting no history of low back pain within the last year. Subjects' ages and anthropometric measures, including weight, height, and knuckle height, were recorded.

2.3 Procedures

2.3.1 Electrode Placement

This study concentrated on APDF profiles of the ES muscle activity during sagittally symmetrical and asymmetrical lifting of loads based on the 1991 NIOSH equation. Surface electrodes were chosen over intramuscular wire electrodes based on their non-invasive application, their day-to-day reliability, and their more general pick- up area (Giroux and Lamontagne, 1990). To clearly determine the extent of ES involvement during lifting tasks, multilevel, bilateral electrode placement was utilized since previous work has revealed functional differences of the paraspinal muscles with respect to frontal and sagittal plane loading (Vink *et al.*, 1988). Six pairs of bipolar, Ag/AgCl surface electrodes were placed bilaterally at vertebral levels T9, L1, and L3 parallel to the spine (Lafortune *et al.*, 1988). The spinous processes of the lumbar vertebrae were located by palpation and marked with indelible ink. Electrode pairs were placed 4 cm lateral to the spinous process at the T9 level, 9 cm lateral to the spinous process at the L1 level and 3 cm lateral to the spinous process at the L3 level (Figure 1). Prior to electrode placement, the specified sites were shaved with a razor and scrubbed with rubbing alcohol to reduce skin resistance as much as possible. An ohmmeter was used to test skin resistance. Electrodes were reattached if a resistance greater than 5 kΩ was recorded (cf., Winter, 1988).

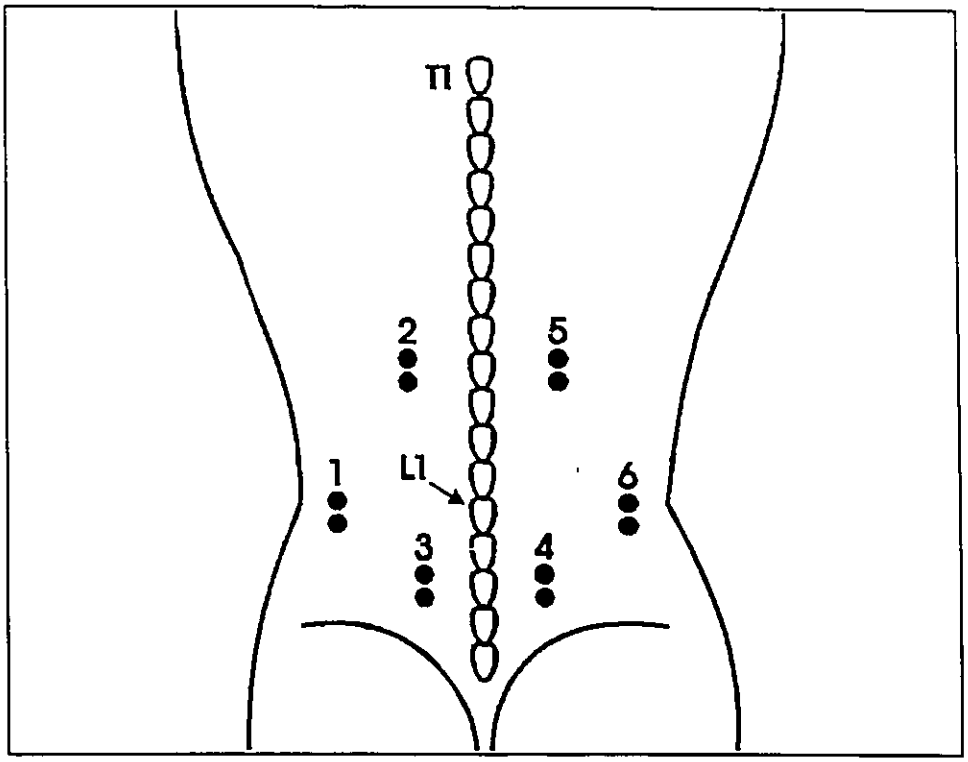


FIGURE 1. Bilateral electrode placement of the ES. Electrodes are placed 4 cm lateral to the midline at T9, 9 cm lateral to the midline at L1, and 3 cm lateral to the midline at L3.

2.3.2 Maximum Voluntary Contraction

All of the trial APDFs were normalized to percent ES MVC values taken during the same lifting session. The MVC strategy employed a psychophysically determined maximum deadlift of an Olympic weightlifting bar and weight plates. The deadlift provided an accurate dynamic simulation of the task measured in this study while maintaining the high safety level necessary when loads of this magnitude are placed on the musculoskeletal system. Prior to any lifting, subjects followed a stretching routine for the trunk and legs to reduce the chance of injury, and were trained in proper deadlift technique. A dual overhand grip was used with the hands placed on the bar slightly wider than shoulder width. For the starting position, the ankle, knee and hip joints were bent, a natural lordodic curve was maintained in the lumbar spine, and the arms were kept relatively straight with the elbows pointed outwards. To perform the lift, the subjects were instructed to apply some lifting force to the bar to preload the musculoskeletal system, and then perform the lift finishing in a standing position holding the bar at knuckle height. The lift was initiated with the legs and incorporated trunk extension after the bar was raised superior to the knees. Throughout the lifting tasks, there was an emphasis made on maintaining a natural lordodic curve of the lumbar spine. Subjects started with a lifting weight of 70 kg, equal to 70% of the maximum isometric forces measured for this squat lift posture (Ruhmann and Schmidtke, 1989). After each deadlift, 5-10 kg were added to the barbell until the subject expressed that his maximum was reached, or until the researcher determined that lifting technique was being compromised.

EMG activity was collected for each lift during the positive work phase. EMG collection started just prior to the lift and finished after the weight was raised with the subject standing erect. Komi (1973) has shown that maximal EMG does not depend on the length / tension / speed of contraction relationship of the the contractile element. The present study dealt with determining the state of activation of the contractile element as it related to an MVC, and did not warrant the measurement of muscle length or speed of contraction. EMG activation was compared relatively across conditions under the assumption that significantly greater contractile element activation values related to significantly greater force generation of the muscle measured.

2.3.3 Lifting Trials

Two different lifting factor variables were tested, the horizontal distance (HF) and the angle of asymmetry (AF). The remaining four NIOSH lifting factors, coupling (CF), vertical location of the hands from the floor (VF), vertical travel distance (DF), and frequency of lifting (FF) were held constant for each subject. The FF was optimized to negate muscle fatigue effects and prevent distortion of the APDF profiles to higher %MVC levels (Jonsson, 1978). Eight trials were performed for five conditions of the HF and AF variables (Table 2). Both variables included a common condition whereby the standard lifting location (SLL) horizontal distance of 25 cm and 0° of asymmetry was used. The HF condition included additional lifts where the HF distances were 44 cm and 63 cm. The AF conditions were varied to angles of

TABLE 2. Values substituted into the 1991 NIOSH equation for the five lifting conditions.

	Condition					
	Horizontal variable			Asymmetry variable		
Lifting Variable	1	2	3	1	4	5
HF (cm)	25	44	63	25	25	25
VF (cm)	Set at 20 for all conditions.					
DF (cm)	Determined by subject knuckle height.					
AF (deg)	0	0	0	0	45	90
CF	Rated good (1.0) for all conditions.					
FF (lifts/min)	Set at 1.0 for all conditions.					

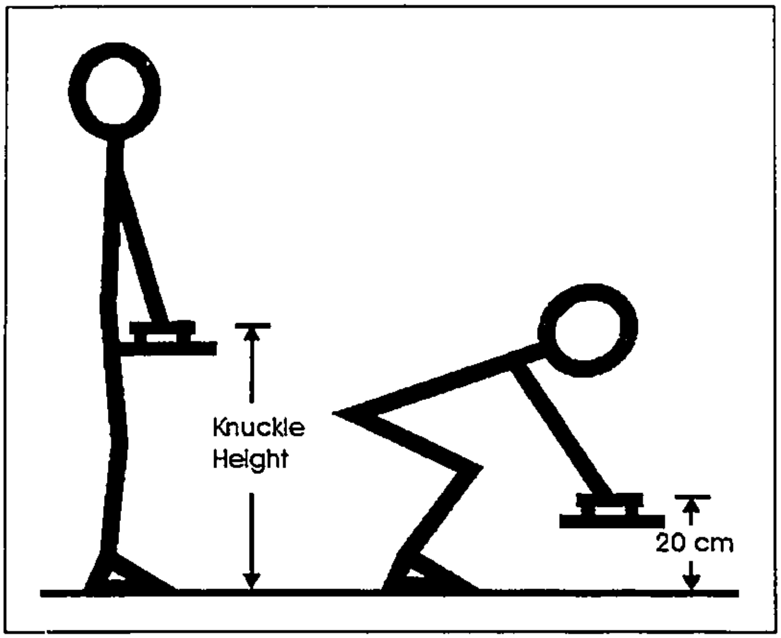


FIGURE 2. DF dimensions for the origin and destination of the load The measurements were taken from ground level to the handle of the tray.

45° and 90° from the sagittal plane. During the EMG recording of the lifting tasks, each subject maintained lumbar lordosis to promote the recruitment of the ES muscles while at the same time minimizing the contribution of vertebral ligaments and facet joints (McGill and Norman, 1986; Potvin *et al.*, 1991). Each lift required the subject to start from a comfortable standing position, reach down and grasp the load, lift the load to knuckle height and assume a comfortable standing posture while holding the load in front of the body (Figure 2). The instructions provided for the lifting trials were as follows:

1. Maintain a "flatback" at all times.
2. Lift with a smooth motion at a natural speed.

3. Grasp and lift the load equally with both hands.
4. Do not slide the load along the raised platform's surface.
5. Maintain a consistent lifting motion for all the lifts.

A total of 40 lifts, 8 lifts per condition, were performed by each subject. The lifts were ordered randomly to counteract learning or fatigue effects. The condition loads were calculated according to the 1991 NIOSH equation:

$$RWL = 23 \times HF \times VF \times DF \times AF \times FF \times CF$$

Where:

RWL= recommended weight limit.

HF = horizontal distance of hands from midpoint between the ankles. Measure at the origin and destination of the lift (cm).

VF = vertical distance of the hands from the floor. Measure at the origin and destination of the lift (cm).

DF = vertical travel distance between the origin and the destination of the lift (cm).

AF = angle of asymmetry -angular displacement of the load from the sagittal plane. Measure at the origin and destination of the lift (degrees).

FF = average frequency rate of lifting measured in lifts/min.. Duration is defined to be: ≤ 1 h; ≤ 2 h; or ≤ 8 h assuming appropriate recovery allowances (Waters *et al.*, 1993).

CF = coupling between the load and the worker's hands (Waters *et al.*, 1993).

The lifting trays were loaded with weight plates to total a combined load equal to the RWL and produce a lifting index (LI) equal to 1. A LI greater than 1 is believed to put a certain fraction of workers at an increased risk of sustaining a work-related injury (Waters *et al.*, 1993). The LI is simply:

$$\text{LI} = \frac{\text{Load Weight}}{\text{Recommended Weight Limit (RWL)}}$$

Where:

Load Weight = weight of the object being lifted (kg).

2.3.4 Lifting Apparatus

The lifting apparatus consisted of a force platform (Kistler), a raised surface that outlined the lifting dimensions of the five lifting conditions to the NIOSH defined SLL, and five identical lifting trays that supported the NIOSH calculated loads (Figure 3). The SLL represents a three-dimensional lifting reference point with its origin 75 cm above the floor and 25 cm forward of a point midway between the ankles in the mid-sagittal plane. The raised surface was designed to isolate the subject on the force platform (Kistler) and allow the positioning of the load to reach the minimum HF distance. The height of the raised surface, in conjunction with the lifting tray dimensions, maintained the VF position of the hands at 20 cm.

The DF was determined for each subject by measuring the knuckle height while the tray was held in front of the body during quiet, steady standing. The DF of the lifts used, were consistent with Snook's and Ciriello's (1991) floor-to- knuckle height lift definition. The lifting trays were designed with handles 45 cm apart which corresponded to the criteria of a "good" CF rating outlined by Putz-Anderson and Waters (1991). The trays were loaded with weight plates according to the NIOSH calculated RWL for the five lifting conditions

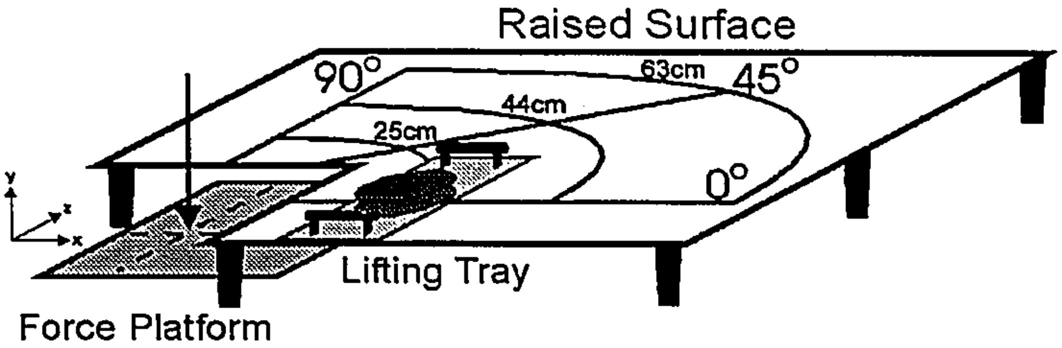


FIGURE 3: Experimental apparatus showing the lifting condition starting locations relative to the force platform. The arrow indicates the point midway between the subject's ankles.

2.4 Data Reduction

All raw EMG signals from the six electrode sites were differentially amplified (Grass Instruments: gain of 2000, input impedance 10 MΩ, 80 dB CMRR at 60 Hz) and band-passed filtered (10-1000 Hz) to remove low and high frequency artifacts (Figure 4). EMG and force plate (Kistler) data were synchronously sampled for a period of 5 seconds for each lift and A/D converted at 1050 Hz. All signals were collected by BIOWARE 2.0 software (Kistler) using a 386 Kinteak computer and stored on floppy disk. The raw EMG signals were full-wave rectified and filtered at a cut-off frequency of 4 Hz to obtain a linear envelope of the EMG signal. Cut-off values from 3 Hz (McGill, 1991b) to 6 Hz (Lafortune et al., 1988) have been reported for linear enveloped EMG signals of the ES during lifting movements. A 4 Hz cutoff was used as the linear enveloped EMG profiles were adequately smoothed without appreciable attenuation of the peak amplitudes. A 6 Hz cutoff frequency produced a noisy profile which was not ideal for averaging trials.

The vertical force plate profiles (Fy) were used to define the lifting sequence of the condition trials. The lift initiation was indicated by a marked decrease in the Fy force curve and the lift termination occurred when the Fy force curve stabilized at body weight plus load weight. For each pair of electrodes, subject ensemble averages and standard deviations (Yang and Winter, 1984) of all trials, for the five conditions were created. This procedure normalized the time-base of the lifts from 0 to 100% lift duration. The peak values for each pair of electrodes from the linear envelope EMG maximum deadlift trials were used to normalize each subject's ensemble averaged EMG curves to a %MVC scale. Condition averages for each electrode site were determined by ensemble averaging across subjects.

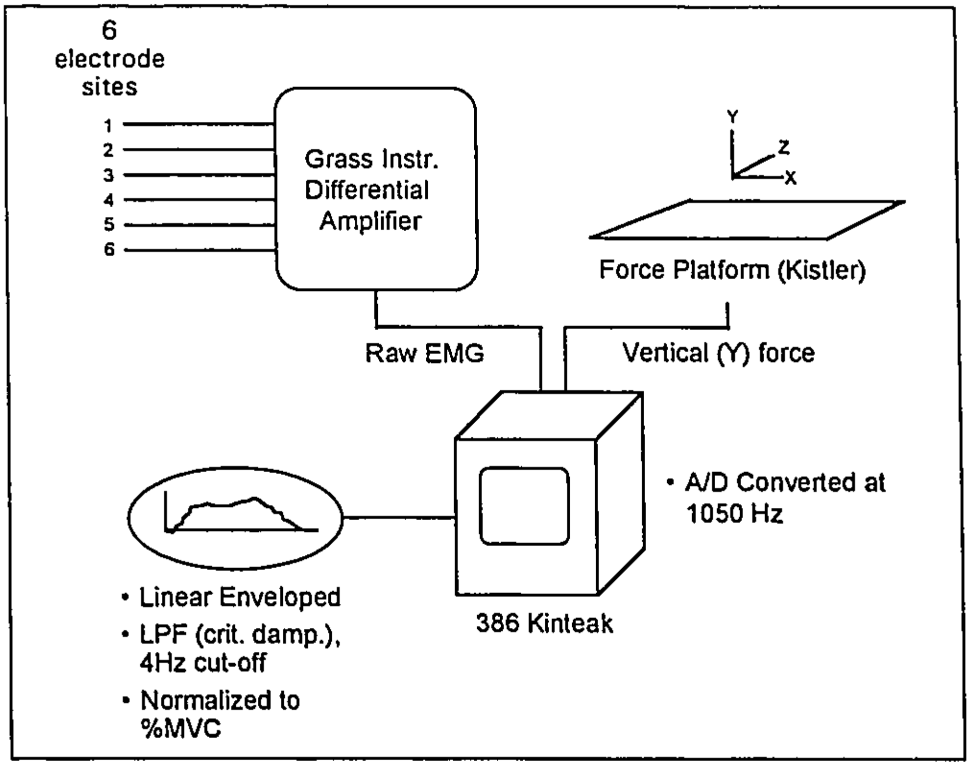
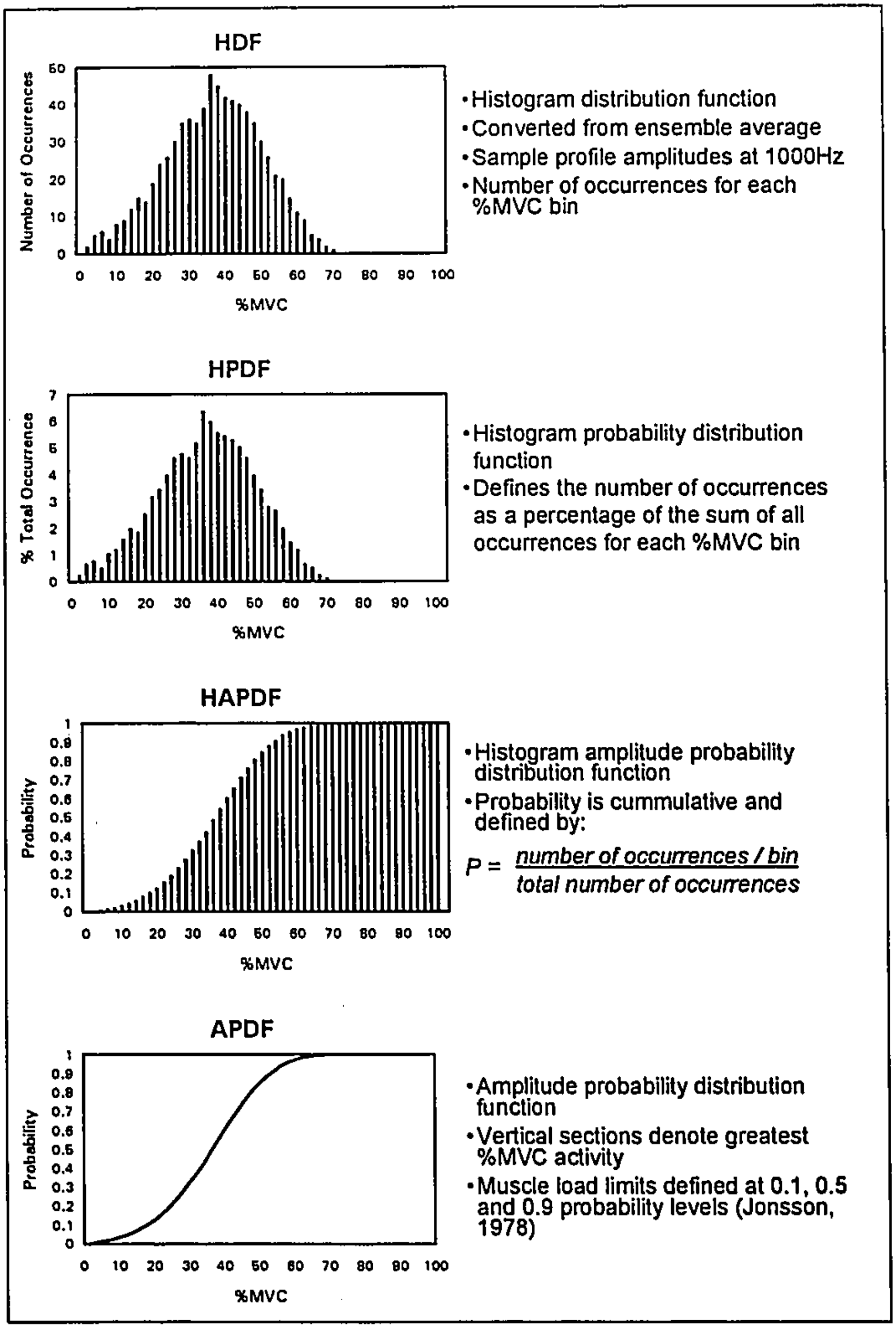


FIGURE 4: Schematic of the data collection apparatus. The normalized, LE EMG profiles were generated for 6 electrode sites for 8 lifts/5 lifting conditions/10 subjects.

Figure 5 shows how the APDF, can produce the muscular load profile based on the EMG amplitude values collected during a specific task for each electrode site (Jonsson, 1982). The condition ensemble averaged EMG curves were converted to APDFs by sampling the amplitude values 1000 times and counting the number of occurrences for each 1% MVC increment from 0 to 100% MVC (Jonsson, 1976; Jonsson, 1978). The sum of all amplitude samples for the APDF was given a probability value of 1. The condition APDF curves thus showed the cummulative %MVC values from 0 to 100% MVC as a probability



- Histogram distribution function
- Converted from ensemble average
- Sample profile amplitudes at 1000Hz
- Number of occurrences for each %MVC bin

- Histogram probability distribution function
- Defines the number of occurrences as a percentage of the sum of all occurrences for each %MVC bin

- Histogram amplitude probability distribution function
- Probability is cummulative and defined by:

$$P = \frac{\textit{number of occurrences / bin}}{\textit{total number of occurrences}}$$

- Amplitude probability distribution function
- Vertical sections denote greatest %MVC activity
- Muscle load limits defined at 0.1, 0.5 and 0.9 probability levels (Jonsson, 1978)

FIGURE 5: Conversion of normalized EMG amplitudes to an APDF.

function from 0 to 1. Thus, more vertical sections of the APDF denoted increased occurrences at those %MVC values while more horizontal sections of the APDF denoted decreased occurrences at those %MVC values.

2.5 Experimental Design and Data Analysis

The %MVC values were taken from the condition APDF curves at probability levels of 0.1, 0.5 and 0.9 to determine the static, median and peak muscular load level, respectively. These averages were compared to Jonsson's (1982) muscular load limit values. A mixed model, repeated measures, three-way ANOVA tested for significant ($p < 0.01$) main effects of the %MVC means for conditions, electrode placements and APDF probability levels.

3.0 Results

3.1 Descriptive Statistics

All ten subjects completed 40 lifts, 8 for each lifting condition, in random order without exception. There were no complaints of physical discomfort during the data collection and all subjects completed the entire collection period within the expected duration of 2.5 hours. Eight of the 10 subjects psychosomatically determined their maximum deadlift limit while 2 subjects were prevented from increasing their maximum deadlift limit by the researcher due to faltering lifting technique. All subjects were visibly exerted during the maximum trials but none showed any signs of apprehension when lifting loads of these magnitudes. There were no reports or complaints of fatigue during the submaximal lifting trials and subjects relaxed the ES muscle between every lifting trial while monitoring muscle activity levels on an oscilloscope.

Descriptive statistics for the subject pool are in Table 3. Descriptive statistics for condition loads are shown in Table 4. These loads were calculated with values from

TABLE 3. Descriptive statistics for subjects and maximum deadlift weight.

	Subject Characteristics				**Maximum**
Descriptive Statistics	age (yrs)	weight (kg)	height (cm)	knuckle ht. (cm)	**Deadlift (kg)**
Average	26.2	79.0	177.3	83.1	116.1
Standard Dev.	4.0	7.7	6.3	3.9	19.4
Maximum	33.0	91.8	186.0	91.0	142.9
Minimum	21.0	64.5	170.0	78.0	93.0

TABLE 4. Descriptive statistics for the condition loads calculated with the 1991 NIOSH equation.

Descriptive Statistics	Condition 1	2	3	4	5
Mean (kg)	17.1	9.7	6.8	14.7	12.2
Standard Deviation (kg)	0.1	0.1	0.0	0.1	0.1
Maximum (kg)	17.2	9.8	6.8	14.8	12.3
Minimum (kg)	17.0	9.6	6.7	14.5	12.1

Table 2 as inputs to the six lifting factors. The DF variable was the only value to vary between subjects and hence, affect the condition loads between subjects. As can be seen from Table 4, condition loads remained almost constant across subjects despite a DF range of 13.0 cm. Condition lifting duration was defined as the time taken to reach down from a standing position and lift the load to a standing position. Despite the large range of lift dimensions between the 5 conditions, the mean lift duration remained relatively constant across conditions (Table 5). A one-way analysis of variance showed no significant differences in lift duration between conditions ($p < 0.05$, df = 4). Similar lift durations are important to the interpretation of the APDF. If the durations were not similar, it could be argued that differences between APDF curves were the result of differences in the sampling period.

TABLE 5. Descriptive statistics for condition lifting duration.

Descriptive Statistics	Condition 1	2	3	4	5
Mean (sec.)	3.1	3.0	3.0	3.1	3.2
Standard Deviation (sec.)	0.4	0.3	0.3	0.3	0.3

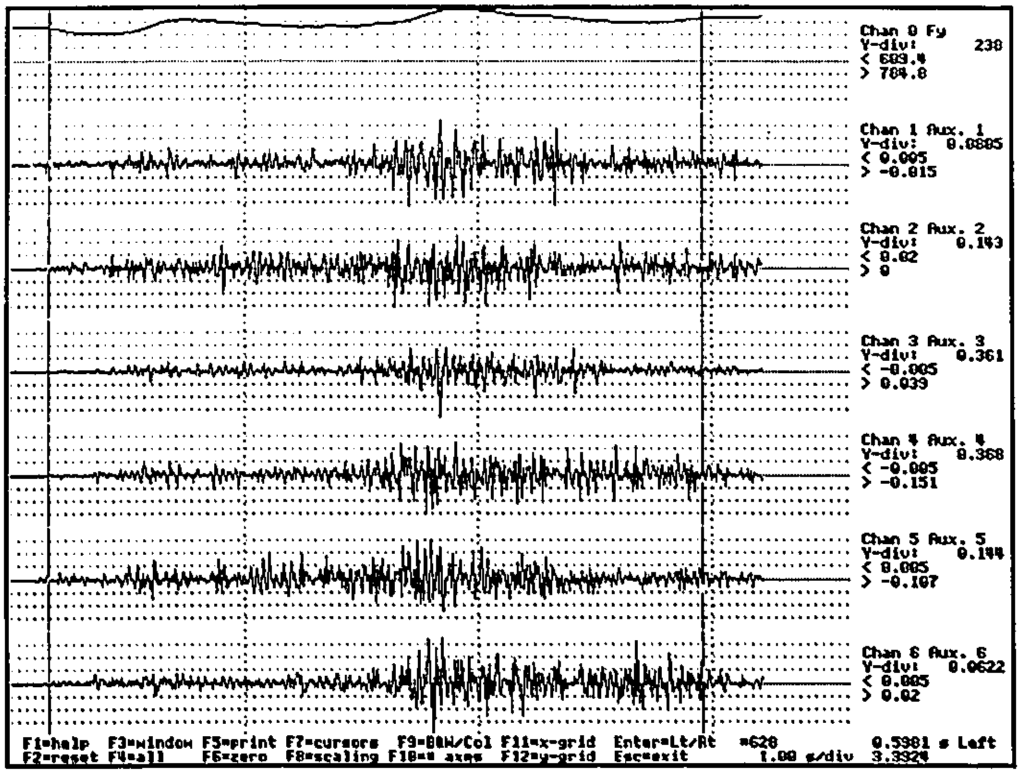


FIGURE 6: Fy force curve and RAW EMG signals (μV) for a lifting trial. The right and left vertical lines show how the force curve was used to window the lifting sequence. Increased EMG signal after the lift is a function of the load the subject must hold after the completion of the lift. Each signal is scaled to its maximum value.

3.2 Electromyography

Figures 6 and 7 are representative of the EMG signals of the six electrode sites and the associated vertical force. These figures depict a typical single trial showing raw EMGs in Figure 6 and LE EMG signals in Figure 7. The lifting sequence is windowed and shown by two vertical lines that define the initiation and termination of data selection.

Figures 8 through 12 display the condition ensemble average of the LE EMG, ± 1 standard deviation, for the six electrode sites. The curves were normalized to 100% MVC

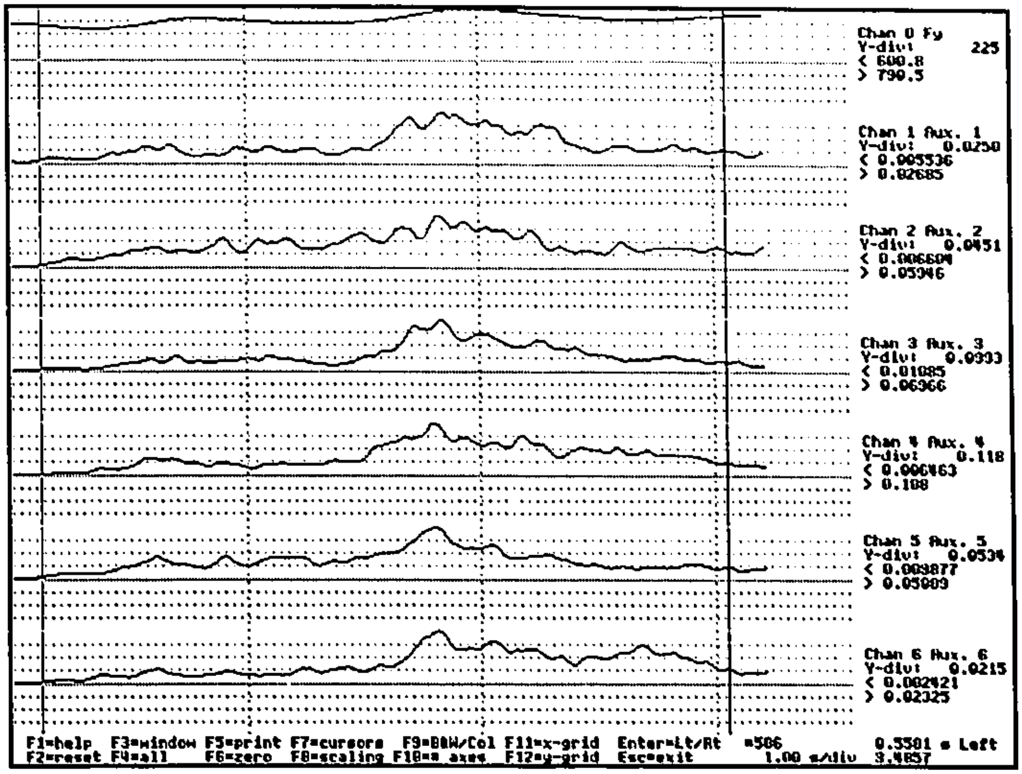


FIGURE 7: Fy force curve and LE EMG signals (μV) for a lifting trial. LE EMGs were produced using a 4Hz critically damped, low-pass filter. Vertical lines show the window of the lifting sequence. Each signal is scaled to its maximum value.

on the ordinal axis and 100% lifting duration on the abscissa. From the LE EMGs, two distinct peaks can be seen, a smaller peak occurring from 0, to approximately 40% lift duration and a larger peak from approximately 40% to 100% lift duration. The first peak corresponds to the reaching down phase of the lift duration and the second peak results from the muscular effort to lift the load. The ratio of reach to effort durations shifts to a longer reach/shorter effort duration by 5% lift duration as the HF increases (Figures 8 to 10). This relationship would require greater acceleration of the load during the initial stage of the effort phase to compensate for a decreased time duration. Indeed, lifts with a

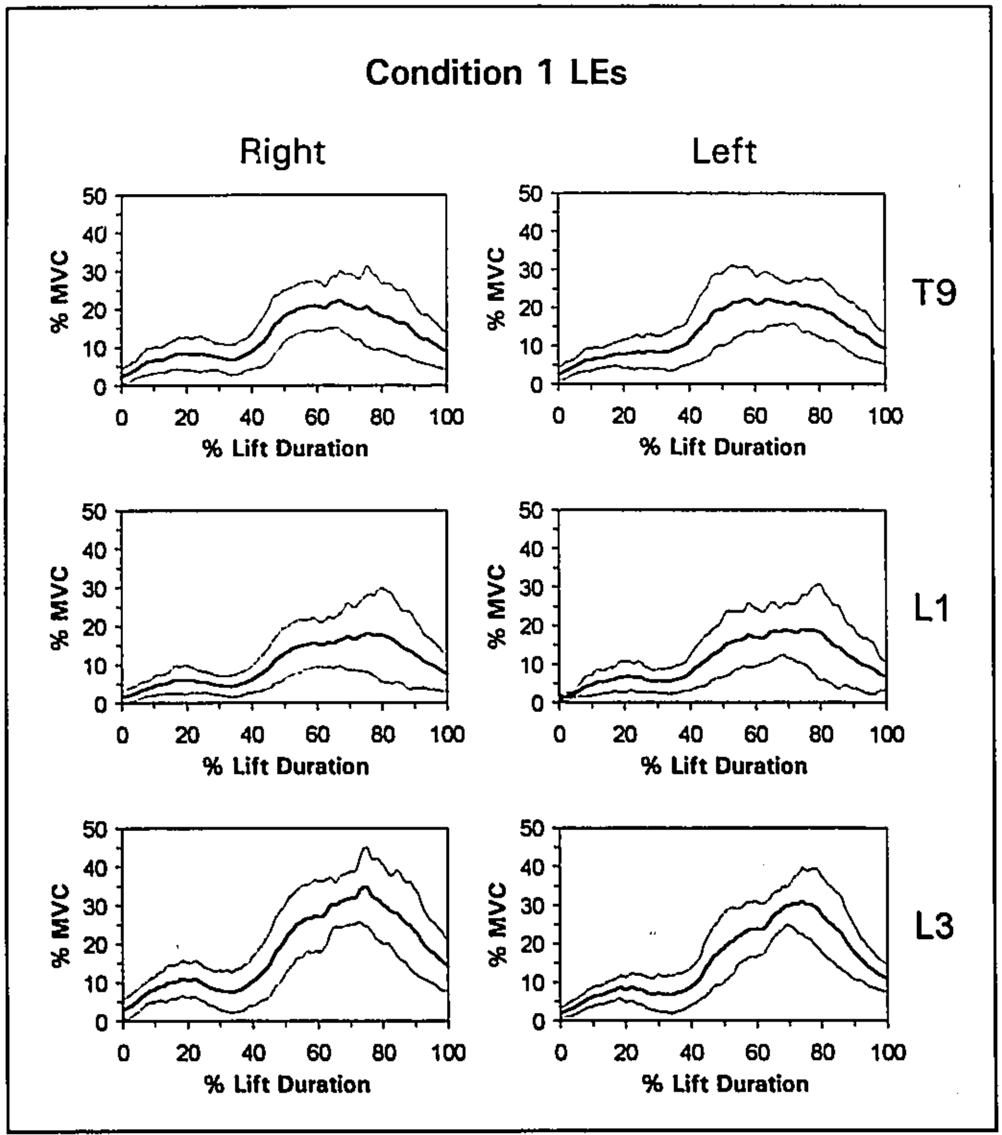


FIGURE 8. Linear enveloped EMG curves for the six electrode sites and their respective standard deviations. Amplitudes were normalized to 100% MVC and the sequence was normalized to 100% lift duration. The curves are arranged as right and left ES and paired for each of the three vertebral levels measured. All profiles were created by averaging 10 subjects, 8 trials per subject. The lifting dimension variables for these LE EMG results were: HF=25 cm and AF=0°.

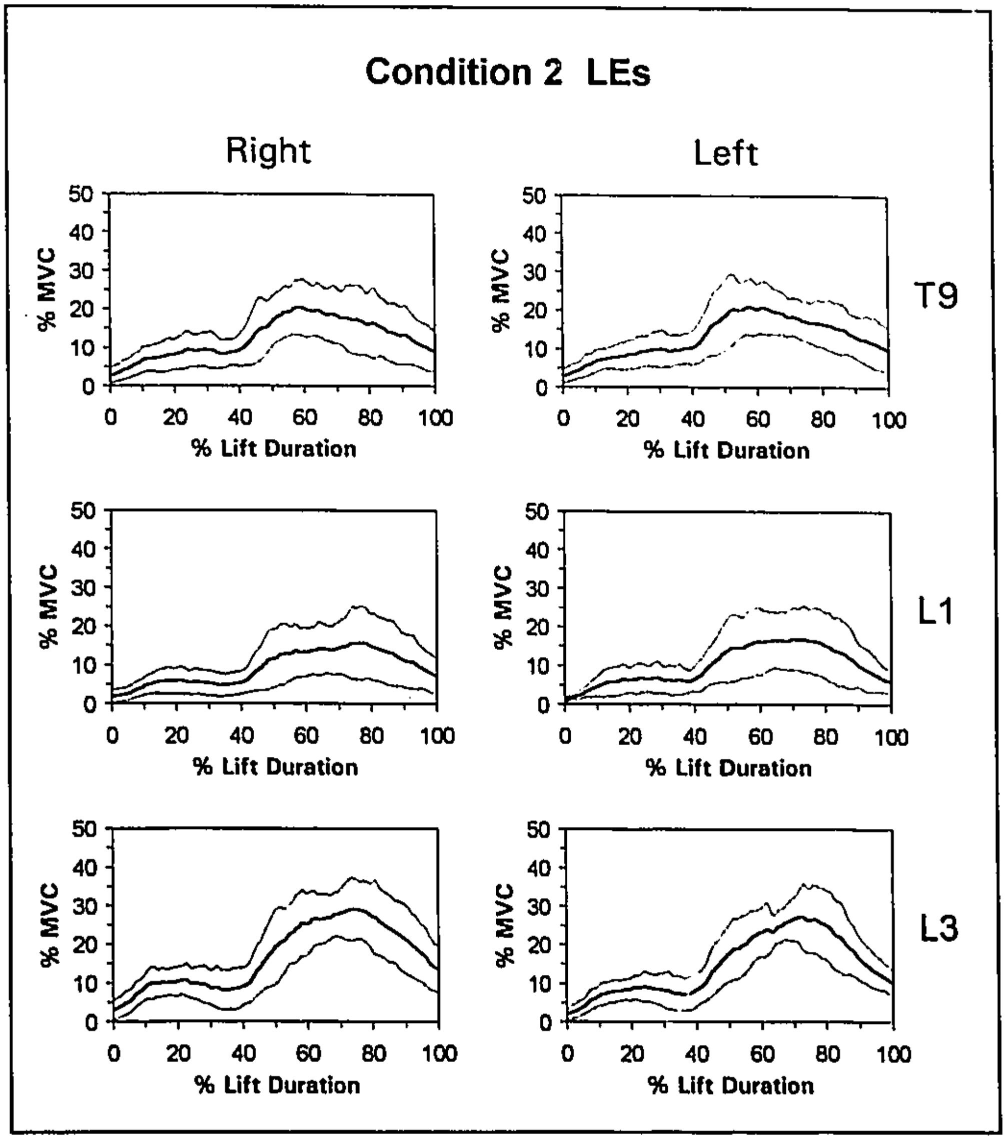


FIGURE 9. Linear enveloped EMG curves for the six electrode sites and their respective standard deviations. Amplitudes were normalized to 100% MVC and the sequence was normalized to 100% lift duration. The curves are arranged as right and left ES and paired for each of the three vertebral levels measured. All profiles were created by averaging 10 subjects, 8 trials per subject. The lifting dimension variables for these LE EMG results were: HF=44 cm and AF=0°.

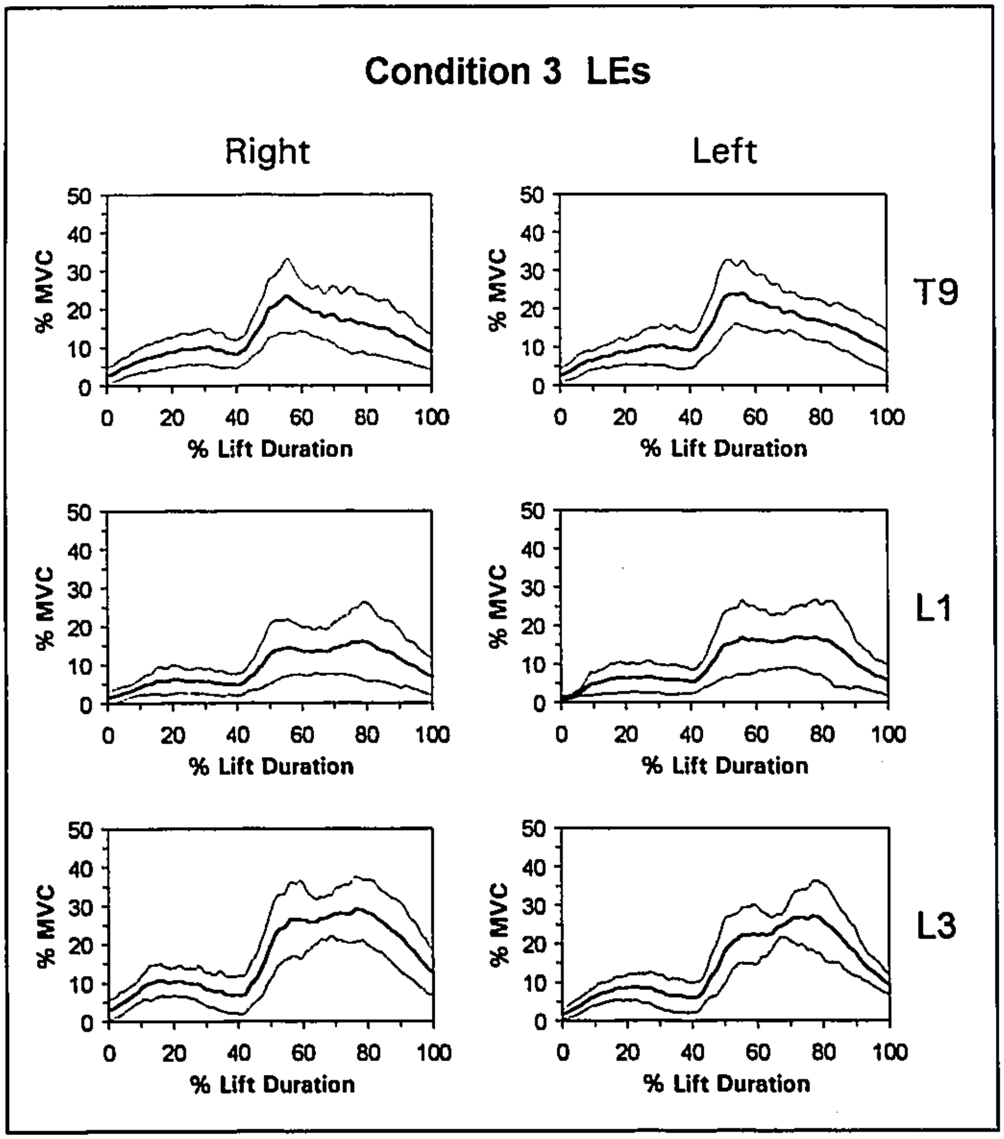


FIGURE 10. Linear enveloped EMG curves for the six electrode sites and their respective standard deviations. Amplitudes were normalized to 100% MVC and the sequence was normalized to 100% lift duration. The curves are arranged as right and left ES and paired for each of the three vertebral levels measured. All profiles were created by averaging 10 subjects, 8 trials per subject. The lifting dimension variables for these LE EMG results were: HF=63 cm and AF=0°.

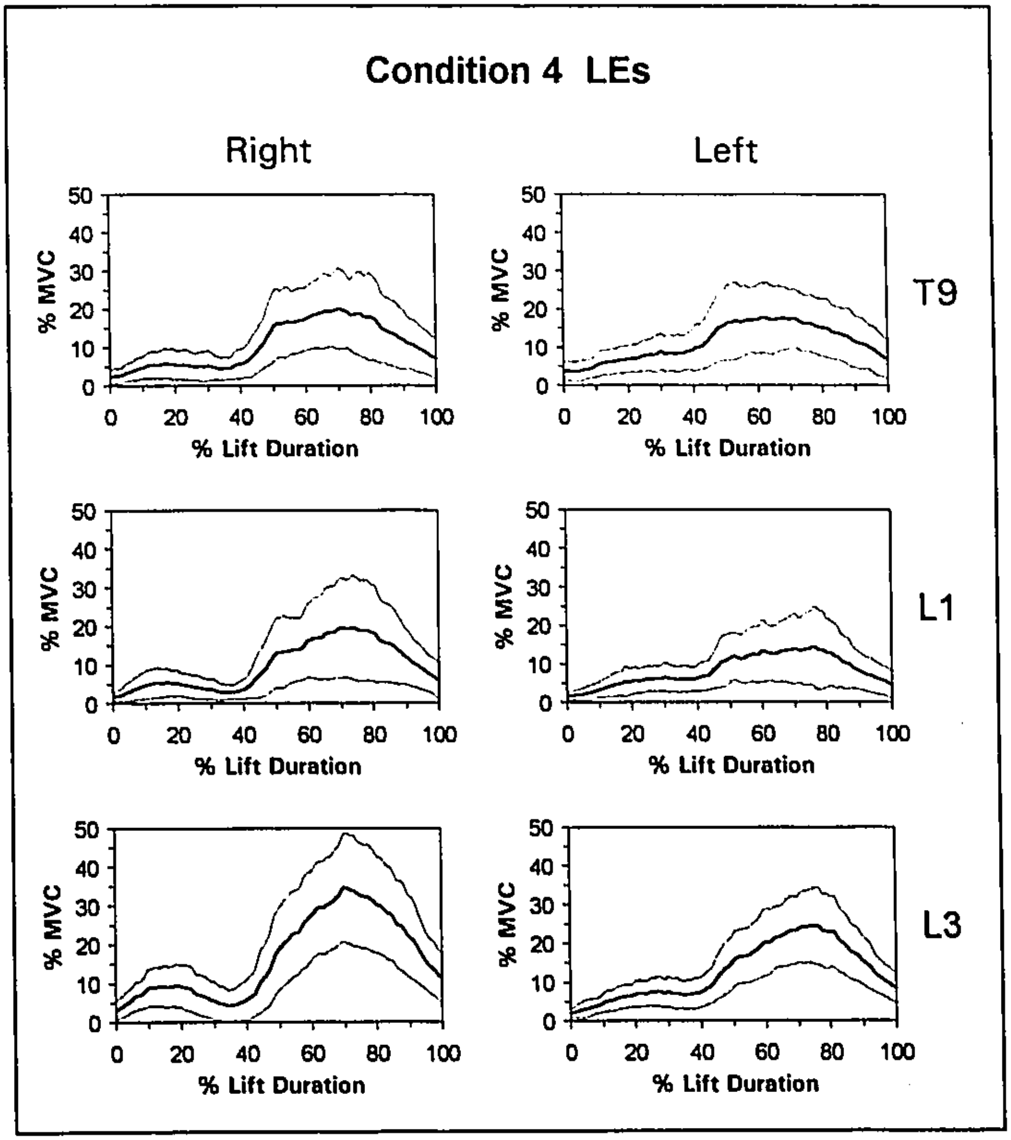


FIGURE 11. Linear enveloped EMG curves for the six electrode sites and their respective standard deviations. Amplitudes were normalized to 100% MVC and the sequence was normalized to 100% lift duration. The curves are arranged as right and left ES and paired for each of the three vertebral levels measured. All profiles were created by averaging 10 subjects, 8 trials per subject. The lifting dimension variables for these LE EMG results were: HF = 25 cm and AF = 45°.

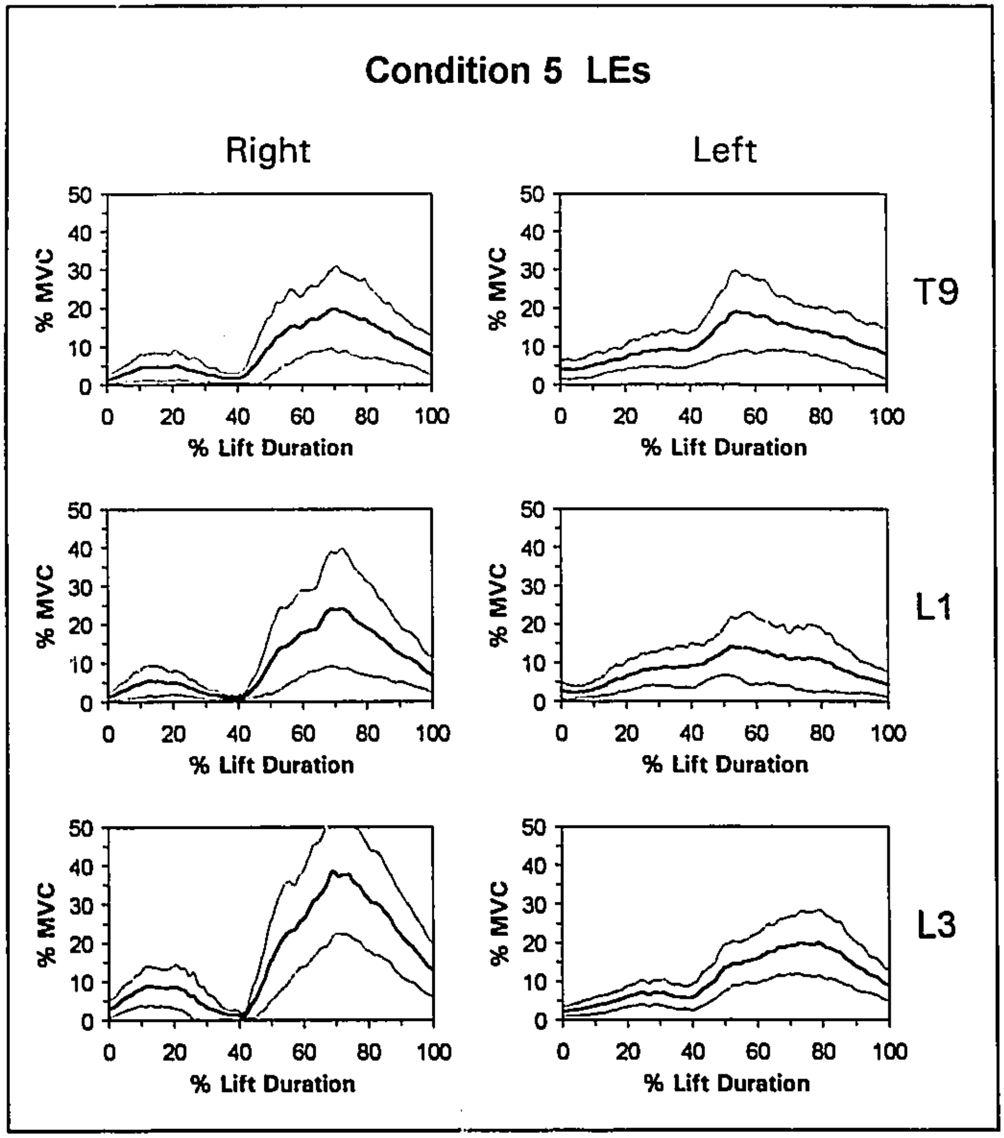


FIGURE 12. Linear enveloped EMG curves for the six electrode sites and their respective standard deviations. Amplitudes were normalized to 100% MVC and the sequence was normalized to 100% lift duration. The curves are arranged as right and left ES and paired for each of the three vertebral levels measured. All profiles were created by averaging 10 subjects, 8 trials per subject. The lifting dimension variables for these LE EMG results were: HF=25 cm and AF=90°.

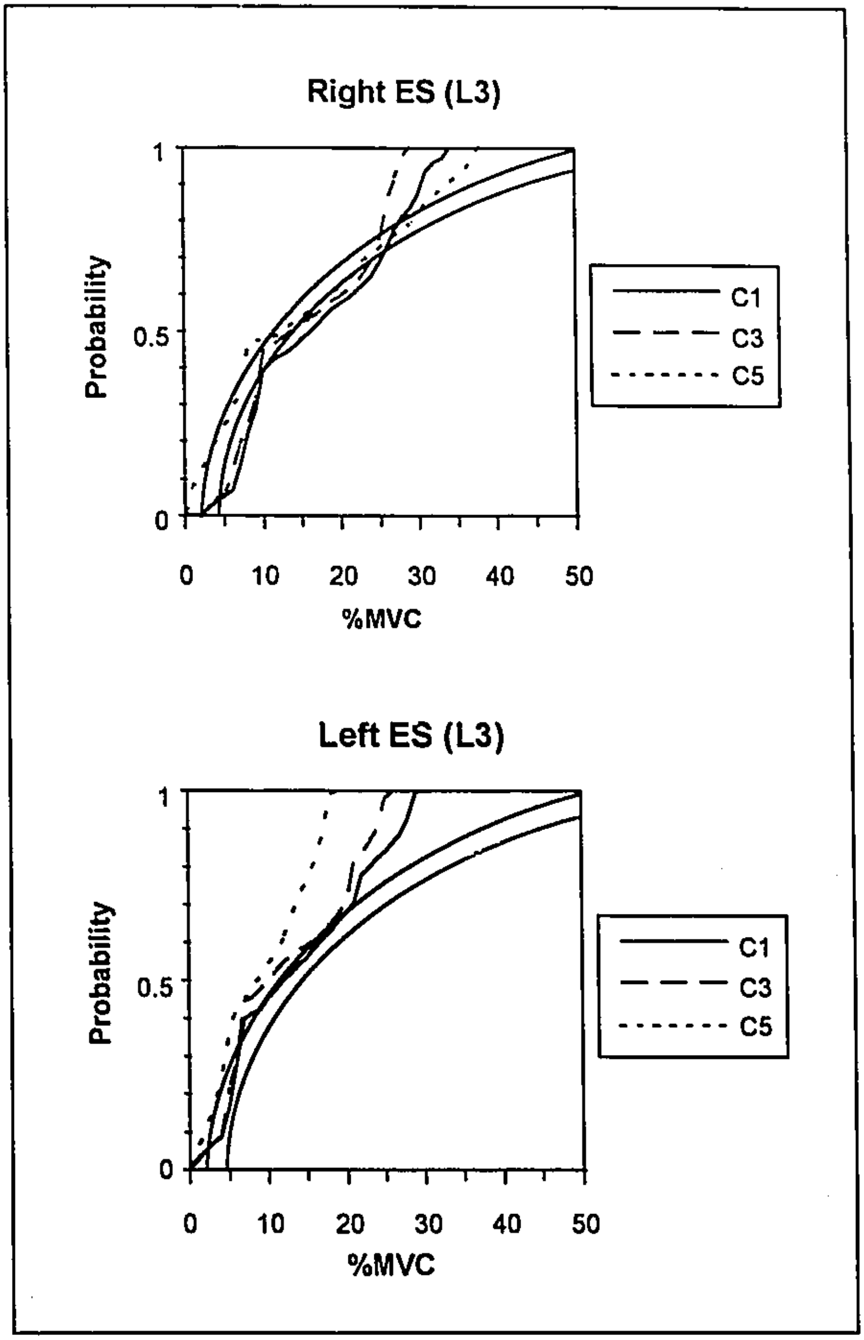


FIGURE 13. APDF curves for the right and left ES(L3) for conditions 1,3 and 5. The arced lines are a representation of Jonsson's (1978) threshold limit APDF that was suggested for muscular endurance for continuous work. APDFs to the right of this zone are considered within acceptable values. APDFs to the left of this zone would indicate that ergonomic correction of the workplace, or task, is required.

63 cm HF were characterized by a sharp rise of the LE EMG signatures, at approximately 42% of lift duration, seen at all six electrode sites (Figure 10). The peak amplitudes across the HF lift conditions remained constant indicating the effectiveness of the NIOSH load adjustment for these lifts. The AF lift conditions (Figures 8, 11 and 12) also show a shift to a longer reach phase and shorter effort phase of the lift duration, for increasing degrees of asymmetry, although it is not as obvious as the HF lift conditions. The distinct pattern of the AF conditions was the increasing asymmetry between the right and left sides. The right ES was contralateral to the load. As the load was displaced asymmetrically to the left of the subject, the right ES became more active during the effort phase. The reaching phase did not appear to show an appreciable change for the right ES but did show decreased activity for the left, ipsilateral, ES under increasing asymmetry. In all cases (Figures 8 to 12), the L3 ES displayed the greatest peak activity.

3.3 Amplitude Probability Distribution Function

Figure 13 highlights the APDF curves. These curves were generated by summing the occurrences of condition LE EMG amplitudes for each %MVC increment from 0 to 100% MVC and plotting those occurrences as a probability function, the total number of occurrences equal to a probability of 1. Conditions 1, 3 and 5 are shown for the right and left L3 ES electrode sites. The arced lines indicate the zone whereby muscular load limits are approaching a level of risk. Any muscular load limits to the right of this zone are unacceptable for prolonged work tasks (Jonsson, 1978, 1982). APDF curves were

TABLE 6. Condition mean %MVC values for the APDF probability levels.

	T9		L1		L3	
Condition	Right	Left	Right	Left	Right	Left
			Probability level 0.1			
1	5.5	5.6	3.5	3.9	5.7	4.6
2	5.9	6.0	3.9	4.1	6.5	5.2
3	5.8	6.0	3.6	4.0	5.9	4.5
4	4.6	6.2	3.2	3.6	4.9	4.4
5	2.4	5.8	1.8	3.5	2.3	3.5
			Probability level 0.5			
1	11.7	12.8	8.8	9.2	15.1	12.5
2	11.3	12.3	8.2	8.6	14.9	12.1
3	11.7	12.5	8.3	8.3	15.1	11.8
4	11.5	12.9	9.1	8.8	14.9	12.3
5	8.5	12.9	8.3	9.6	13.5	11.1
			Probability level 0.9			
1	23.2	24.2	18.9	20.3	34.2	30.9
2	21.4	21.7	16.3	18.2	30.3	28.4
3	21.6	23.2	16.4	18.6	30.0	28.0
4	23.0	21.4	21.0	16.8	35.4	26.4
5	21.6	21.2	23.5	15.9	37.6	21.9

TABLE 7. Condition APDF %MVC values that exceed muscular load limits for probability levels 0.1, 0.5 and 0.9.

	Condition				
Electrode	1	2	3	4	5
Right ES (T9)	.1	.1	.1	-	-
Left ES(T9)	.1	.1	.1	.1	.1
Right ES(L1)	-	-	-	-	-
Left ES(L1)	-	-	-	-	-
Right ES(L3)	.1,.5	.1,.5	.1,.5	.5	-
Left ES(L3)	-	.1	-	-	-

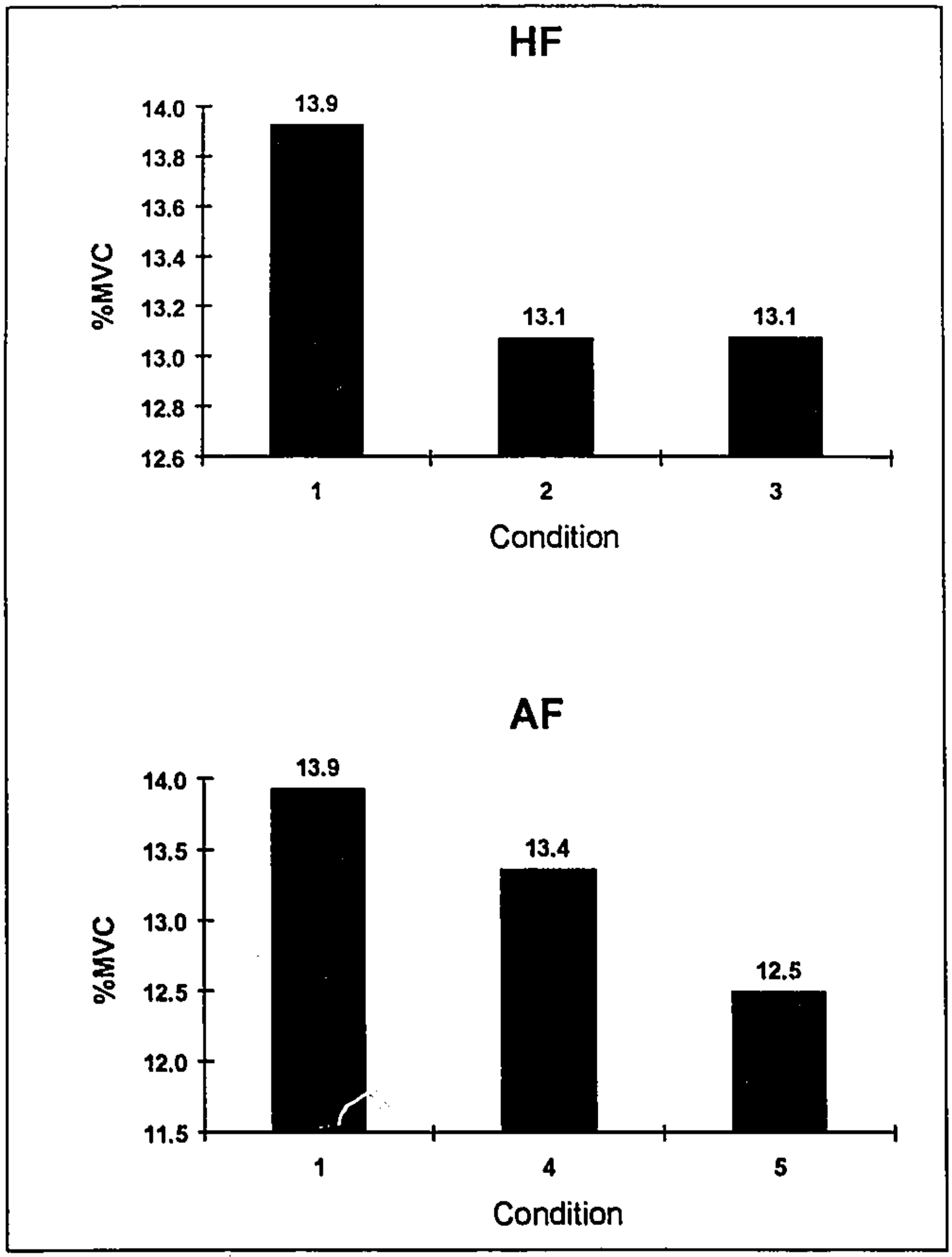


FIGURE 14. Percent MVC means showing the main effect of the condition variable. For purposes of comparison, the means are sorted into the HF and AF.

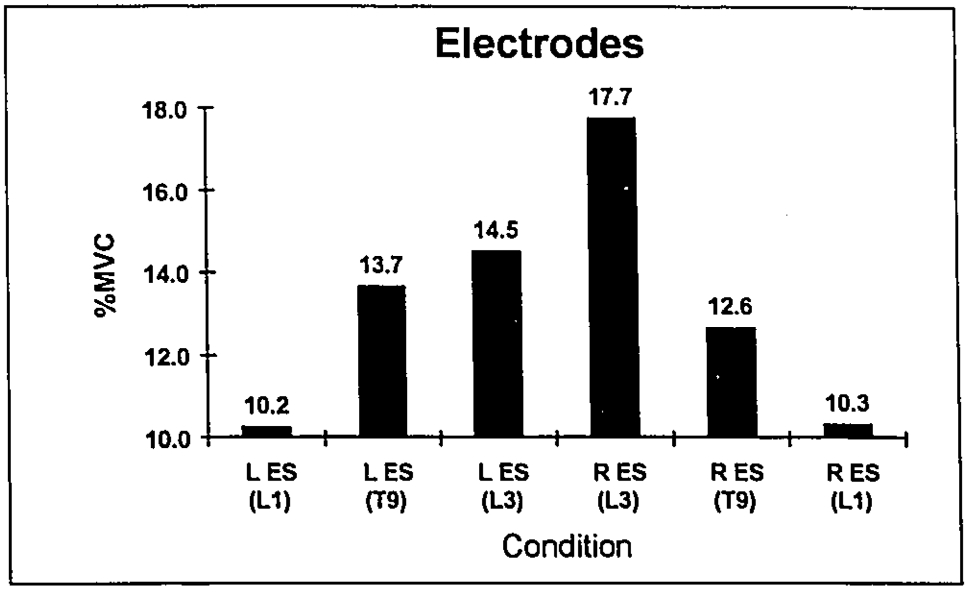


FIGURE 15. Percent MVC means showing the electrode placement main effect.

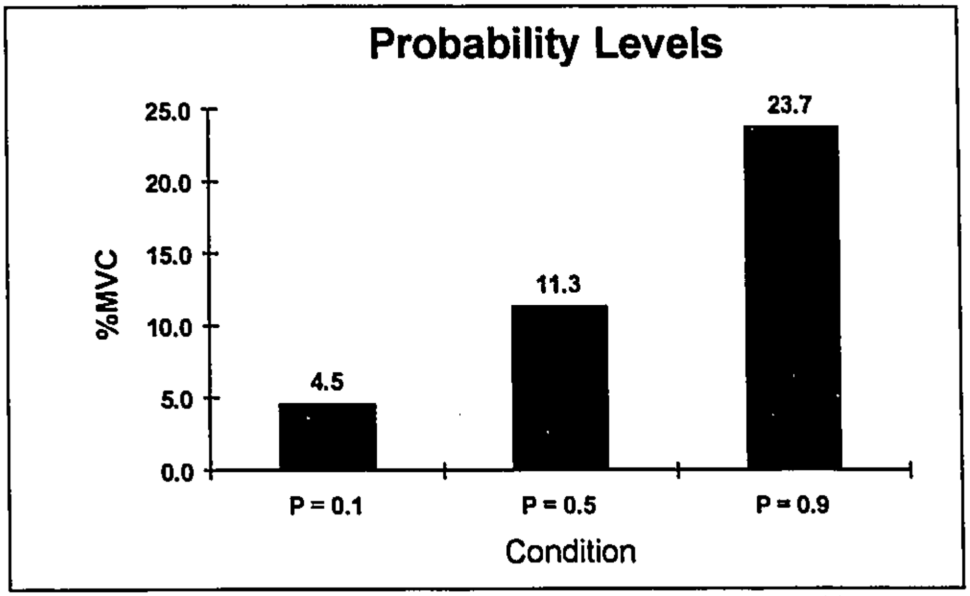


FIGURE 16. Percent MVC means showing the probability level main effect for the condition APDF curves.

generated for all conditions and all electrode sites. The results of the %MVC values associated with the probability levels of 0.1, 0.5 and 0.9 are tabulated in Table 6. Instances where the %MVC values exceed Jonsson's muscular load limits are indicated in Table 7.

3.4 Main Effects

Results of the 3-way, repeated measures, mixed model ANOVA are presented in figures 14 to 16. Main effects were assessed for the three variables: condition, electrode placement, and probability level. In each case, there was a significant ($p < 0.01$) main effect. Figure 14 shows the relationship of the HF conditions and the AF conditions separately. The values on both bar graphs, however, are directly comparable. The %MVC amounts involved the sum of all subject mean %MVCs across all electrodes and all probability levels for each condition. Condition 1 is repeated on both bar graphs. Figure 14 reveals that condition 1 was the most demanding of ES activity levels. The technique of ensemble averaging requires that all EMG samples show an absence of activity at exactly the same cycle time for a value of 0% MVC to register in the condition ensemble average curve. Likewise, a 100% MVC would have to occur for all EMG samples at the same cycle time for a condition ensemble average curve to reach a maximum value.

Figure 15 shows the relationship of the six electrode sites. The %MVC amounts here, involved the sum of all subject mean %MVCs across all conditions and all probability levels for each electrode site. Figure 15 indicates that the lifting scenarios used in this study elicited the most sub-maximal muscle activity from the ES at the L3 level and that the right

L3 ES, contralateral to the load for asymmetrical lifts, reaches the greatest %MVC levels.

Figure 16 illustrates the trend of the APDF probability levels. %MVC levels were taken from the sum of all %MVC means for the 5 conditions and 6 electrode sites for each probability level. The probability level trend increased with increasing probability levels as expected due to the cummulative effect of the APDF

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4.0 Discussion

This study was designed to compare the 1991 NIOSH lifting guidelines with muscular loads of the ES during specific, dynamic, lifting tasks. During this process it was possible to determine whether the NIOSH lifting equation adequately corrected for different lifting conditions through a reduction in the load based on ES EMG. Normalized EMG was used to generate an APDF from which static, median, and peak muscular load limits were interpreted. The APDF results were in general agreement with the 1991 NIOSH equation. Most condition %MVC means were within the threshold limit APDF which defines muscular load limits (Jonsson, 1978). Exceeding these limits would require ergonomic correction to the worksite, or task, due to an increased risk of injury to the worker.

4.1 EMG Normalization

To interpret the APDF curve, it was necessary to convert the EMG data from mV units to a functional measure of the muscles' capacity to contract voluntarily. A maximum deadlift was performed to produce an MVC value for the electrode sites of the ES. Often researchers have normalized EMG data to a maximum voluntary contraction, elicited during an isometric exertion, at a specific joint angle (McGill, 1991b; Stalhammar and Louhevaara, 1992). Large errors can occur under these circumstances as the portion of the muscle within the viewing area of the electrode and the length/tension relationship of the muscle vary at different joint angles. This would result in changes to the MVC value. It has been suggested that an MVC should be sampled under a range of joint angles (McGill, 1991b; Mirka, 1991).

In the present study, MVCs were generated from a dynamic lift, avoiding the problems associated with normalizing dynamic activities with an isometrically derived EMG amplitude at a specific joint angle. Furthermore, the range of joint angles performed during the test contraction were similar to the trial contractions. Using this approach, EMG amplitudes reached 2 mV during the deadlift MVC; whereas Stokes *et al.* (1987) reported levels of 120 μV for isometric MVCs of the same L3 ES electrode location. Yang and Winter (1984) have cautioned the use of normalizing EMG to a 100% MVC stating that it is less reliable than using a submaximal (50% MVC) test contraction due to lower fatiguability, absence of co-contraction, and a linear EMG-force relationship. Stokes *et al.* (1987) reported a bilinear relationship for ES EMG which contained a breakpoint between the two EMG-force representative slopes for isometric contractions at various joint angles. They found, however, that for repeated measures the slope of a single linear relationship was less variable than the slopes of a bilinear relationship. It is not known if their results were applicable to a dynamic task. Due to the slow, controlled nature of the movements in the present study, it was assumed that a linear relationship existed from 0 to 100% MVC, and that the deadlift should be viewed as a functional MVC, rather than an absolute MVC, due to the direct relationship of the test contraction activity to the submaximal trials. The test contraction used in the present study was the best estimate of an MVC.

4.2 Lifting Cycle

The use of the APDF in the interpretation of EMG amplitudes requires clear reporting of the duration of data collection and the events that occur during this duration. This study defined the lift duration as initiating when the subject's centre of mass accelerated downward from a standing position, and terminated when the subject returned to a standing position and the centre of mass stabilized. While other authors have used either start or end points resembling the ones used in the present study to define their lifting duration (Freivalds *et al.*, 1984; Lafortune *et al.*, 1988; Louhevaara *et al.*, 1990; Leskinen *et al.*, 1992), only Potvin *et al.* (1991) used a similar lifting duration definition when investigating stoop lifts. Previous definitions of lift initiation used an upward force applied to the load (Freivalds *et al.*, 1984; Lafortune *et al.*, 1988) or when the load was touched (Potvin *et al.*, 1991). A lift initiation that occurs after the initial upward force production on the load may fail to record potentially important EMG amplitudes. Ensemble averaged linear envelopes from Lafortune *et al.* (1988) showed that near peak EMG amplitudes occurred at the initiation of their lift sequence which omitted large amplitude EMGs. A comparison of their phasic ensemble averaged, LE EMG patterns, to the patterns in the present study, would place the initiation of their lift at approximately 60% of lift duration. Figure 6 shows inactive ES raw EMG prior to the lift initiation and active ES after the lift initiation. This procedure eliminated the chance that large amplitudes would be omitted prior to the lift. Most studies agreed that the termination of the lifts occurs at upright standing while holding the load. At this point, ES EMG remained active, within 5 to 15 %MVC, likely a result of an extensor moment required to support the load.

4.3 Electrode Sites

Results of the 3-way, repeated measures, mixed model ANOVA showed a significant ($p < 0.01$) electrode placement effect (Figure 15). The effect of electrodes was likely a function of amplitude differences rather than phase differences as the patterns appeared similar across electrode sites for all conditions. The sum of %MVC means across conditions and probability levels revealed no consistent pattern for bilateral comparison of electrode pairs. The values of Figure 15 include symmetric (conditions 1, 2 and 3) and asymmetric (conditions 4 and 5) lifts and, thus, an elevated right side %MVC value would be expected as this ES was contralateral to the load for asymmetric lifts. This pattern, however, was only prevalent with the L3 ES. The T9 ES %MVC levels were slightly elevated for the left side and the L1 ES %MVC levels remained bilaterally equal. It was unclear why all electrode pairs did not show a right side bias. Comparison of electrode pairs did reveal that the L3 ES bilateral location produced the greatest overall amount of electrical activity. A sum of the paired electrodes produced 32.2% MVC for the L3 ES, 26.3% MVC for the T9 ES and 20.5% MVC for the L1 ES. These findings support the notion of Lafortune *et al.* (1988) that selecting an electrode site 3 cm lateral to the L3 spinous process could represent the lumbar and thoracic paraspinal muscles for lifting studies that employ a "flatback" lifting technique.

4.4 Condition Effects

4.4.1 Horizontal Variable Factor

Conditions 1, 2 and 3 combined to provide the HF effect (Figure 14). It is clear that the 1991 NIOSH equation adequately adjusted the load for changes in the horizontal distance as the overall EMG dropped or remained constant with an increasing HF. These findings are consistent with Garg (1989) who found no significant difference in ratings of perceived exertion when HF was varied. Although his data were based on the NIOSH 1981 lifting equation, it remains relevant to this study as the HF was not modified when the 1991 NIOSH equation was created, except for an increase of the minimal distance allowed from 15 cm to 25 cm. Garg (1989) concluded that the load was effectively reduced for lifts of increasing horizontal distance. Bilateral differences, in the present study, were not tested for significance, however, a strong correlation (r=0.984) existed between the right and left ES for the symmetrical lifting conditions indicating symmetric muscle activity for symmetric lifts.

4.4.2 Asymmetrical Variable Factor

Conditions 1, 4 and 5 combined to provide the AF effect (Figure 14). Although the sum of %MVC means decreased from 0° asymmetry to 90° asymmetry, the highest peak muscular load levels were recorded from the right L3 ES during the condition 5 lift where the load was lifted from a position 90° to the midline. With a mean load adjustment from 17.1 kg at 0° asymmetry, to 14.7 kg at 45° asymmetry, to 12.2 kg at 90° asymmetry, there occurred a mean peak muscular load increase of 34.2%, 35.4% and 37.6% MVC for the contralateral L3 ES, respectively, and a decrease of 30.9%, 28.0% and 21.9% MVC for the ipsilateral L3 ES, respectively. A similar trend occurred with the L1 ES whereas the T9 ES remained fairly consistent across conditions for the peak muscular load levels. These %MVC results suggested that the 90° asymmetry lift, and to a lesser extent the 45° asymmetry lift, placed a larger portion of the load on the contralateral lumbar ES. This was not an unexpected finding as other authors have reported similar trends (Seroussi and Pope, 1987; Lavender *et al.*, 1992). A simplified consequence of asymmetric lifting is the reliance of a unilateral ES in the production of the lifting moment as the ipsilateral ES became an antagonist rather than an agonist. Vink *et al.* (1992) investigated the effect of flexion, lateroflexion and rotation of the trunk on back strength and found up to a 40% reduction in strength for the condition that combined 30° of flexion, lateroflexion and rotation. They attributed the loss of strength to muscle length, muscle activation and the angular moment of the muscle. McGill (1991b) has shown that during maximum axial torque productions EMG values did not exceed 74% MVC for the T9 ES and 61% MVC for the L3 ES. It is not known if the conditions causing asymmetry of the L3 ES shown in the present study are comparable to McGill's (1991) actions used to generate maximal asymmetric trunk muscle activity. However, under the consideration of the research results of McGill (1991) and Vink *et al.* (1992), perhaps the threshold limit APDF would require modification for asymmetric activity to reflect the inability of the musculature to reach a maximum contraction during this unique movement.

4.5 APDF

The APDF can be used to interpret muscular load from normalized EMG taken during a manual materials handling task (Hagberg and Jonsson, 1975; Jonsson, 1982; Andersson and Ortengren, 1984). The ergonomist then has a tool to help determine the level of risk, to the worker, of sustaining a musculoskeletal injury, which can indicate worksite, or task, adjustment if necessary. The APDF curve contains three levels of muscular load information. The threshold limit APDF values provide a guideline for muscular load limits at the three APDF probability levels. These values are applicable only when an MVC has been taken to normalize the EMG data (Andersson and Ortengren, 1984).

The static load component (probability 0.1) reflects muscular rest and is defined by no more than 10% of the total recording time (Jonsson, 1982). Exceeding the APDF limit threshold at this level indicates continuous activity of the muscle during the work task. Hagberg (1979) discovered a high correlation (Pearson's *r* ranging from 0.974 to 0.989) between the APDF of low level intermittent static contraction and the APDF of the related force signal for the elbow flexor musculature. However, during fatiguing conditions of continuous muscle activity for 5-10 minutes, the EMG APDF profile exceeded the force APDF profile. His study revealed the consequences of muscle fatigue affecting the EMG amplitude of the APDF. Results of the static load component in the present study revealed some lifting conditions that exceeded the APDF limit threshold. Since the recording period was of short duration (approximately 3 seconds) and there was total relaxation between each lift, excessive EMG amplitudes at this probability level were not due to elevated EMG as a result of muscular fatigue. The median load component (probability 0.5) defines the median EMG amplitude during the total recording time and is usually quite similar to the mean EMG level from integrated EMG (Jonsson, 1982). The peak load component (probability 0.9) defines the EMG amplitude where higher amplitudes are allowed for only 10% of the total recording time (Jonsson, 1982). Due to the possibility of normalization error, this value usually provides more relevant information than taking an EMG amplitude at a probability level of 1.

Condition APDFs plotted to the right of the threshold limit APDF are considered to have unacceptably high muscular loads. It is evident from Table 6 that the T9 ES and, to a lesser extent, the L3 ES contained excessive static loads for many lifting conditions. Also, the right L3 ES was excessively loaded at the median level for conditions 1 to 4. Peak muscular loads remained below the APDF threshold limit for all cases. The ergonomic interpretation of these findings must be made under the consideration of the nature of the task. The activity recorded was of short duration and isolated specifically to the defined lifting conditions. The 1991 NIOSH equation assumes that minimal activity is performed between repetitive lifts (Waters *et al.*, 1993). Thus, under application of the APDF where lifts are combined with other manual handling activities, a different APDF profile may result mostly influencing the static and median muscular load levels. Another consideration is the actual margin that the threshold limit APDF was exceeded. The static muscular load was, at most, 1.5% MVC greater than the static load limit, and the median muscular load was 1.1% MVC greater than the median load limit.

With slight ergonomic modifications to the tasks, it would be possible to reduce the excessive muscular load limits to satisfy the threshold limit APDF. Under the guidelines of the APDF muscular load limits, the modified lifts could be performed continuously for a 1 hour work period (Jonsson, 1982). Perhaps, for repetitive, low frequency lifting tasks, the most relevant APDF measure is the peak muscular load during tests where only the lift sequence is recorded. The recorded mean peak muscular loads were, at most, 37.6% MVC with one subject producing as much as 47% MVC. These values were far below the muscular load threshold limit of 70% MVC which suggests that the 1991 NIOSH load constant of 23 kg is rather conservative. Some authors have investigated the maximum permissible load for lifting tasks similar to the tasks in the present study and their results confirm that a 23 kg maximum load is a conservative figure (Freivalds *et al.*, 1984; Garg 1989; Ayoub, 1991; Snook and Ciriello, 1991; Ayoub, 1992). The maximum acceptable loads from these studies ranged from 31 kg (Snook and Ciriello) to 41.5 kg (Ayoub). In the present study, the maximum load lifted was 17.2 kg. Karwowski and Brokaw (1992) have mentioned that perhaps the 1991 NIOSH equation contains a high resolution and is more efficient to identify tasks that would place the worker at an increased risk of sustaining an overexertion injury than the older equation which operated from a maximum load of 40 kg.

5.0 Conclusions

Assuming that the threshold limit APDF is applicable to the ES for continuous work, and that the lifts used in this study were biomechanically limiting, there was agreement of acceptability between the load limits determined using the 1991 NIOSH equation and the muscular loads of the ES recorded during those lifts. Peak muscular loads of the APDF curve suggest that the NIOSH derived RWL is rather conservative. The APDFs obtained from the L1 ES would permit the lifts used in this study to occur continuously in the absence of muscular fatigue. The APDFs of T9 and L3 ES suggest the need to reduce median and static muscular loads. Measurement of the ES musculature provided a more sensitive analysis of various lifting conditions than the 1991 NIOSH equation.

The muscular loads of the ES identified an overall reduction of EMG for increasing HF and AF. This relationship suggests that the 1991 NIOSH equation adequately reduces or maintains the level of stress to the musculoskeletal system, when lifts are varied, through a reduction of the RWL.

It is recommended that all three probability levels (0.1, 0.5 and 0.9) of the APDF be considered in the interpretation of muscular load limit results as they each provide unique information. Also, caution should be used in the placement of electrodes when measuring the ES for lifting tasks. Although the ES may display functional similarities, there can be significant differences of contraction levels between electrode locations.

6.0 REFERENCES

Andersson, G.B.J. and Ortengren, R. (1984) Assessment of back load in assemblyline work using electromyography. *Ergonomics*, **27**, 1157-1168.

Andersson, G.B.J., Ortengren, R. and Nachemson, A. (1977) Intradiskal pressure, intra-abdominal pressure and myoelectric back muscle activity related to posture and loading. *Clinical Orthopaedics and Related Research*, **129**, 156-164.

Ayoub, M.M. (1991) Determining permissible lifting loads: an approach. *Proceedings of the Human Factors Society 35th Annual Meeting*, 825-829.

Ayoub, M.M. (1992) Problems and solutions in manual materials handling: the state of the art. *Ergonomics*, **35**, 713-728.

Ayoub, M.M., Chaffin, D.B., Drury, C.G., Herrin, G.B., Kroemer, A.R., Lind, A.R. and Troup, J.D.G. (1982) Work practices guide for manual lifting. *National Institute for Occupational Safety and Health*, Cincinnati, Ohio.

Bean, J.C., Chaffin, D.B. and Schultz, A.B. (1988) Biomechanical model calculation of muscle contraction forces: a double linear programming method. *Journal of Biomechanics*, **21**, 59-66.

Buckle, P.W., Stubbs, D.A., Randle, I.P.M. and Nicholson, A.S. (1992) Limitations in the application of materials handling guidelines. *Ergonomics*, **35**, 955-964.

Chaffin, D.B. (1969) A computerized biomechanical model: development of and use in studying gross body actions. *Journal of Biomechanics*, **2**, 429-441.

Chaffin, D.B. and Andersson, G.B.J. (1991) *Occupational Biomechanics (2nd Edition)*. John Wiley & Sons, Inc, Toronto.

Freivalds A., Chaffin, D.B., Garg, A. and Lee, K.S. (1984) A dynamic biomechanical evaluation of lifting maximum acceptable loads. *Journal of Biomechanics*, **17**, 251-262.

Garg, A. (1989) An evaluation of the NIOSH guidelines for manual lifting, with special reference to horizontal distance. *American Industrial Hygene Association Journal*, **50**, 157-164.

Giroux, B. and Lamontagne, M. (1990) Comparisons between surface electrodes and intramuscular wire electrodes in isometric and dynamic conditions. *Electromyography and Clinical Neurophysiology*, **30**, 397-405.

Hagberg, M. (1979) The amplitude distribution of surface EMG in static and intermittent static muscular performance. *European Journal of Applied Physiology*, **40**, 265-272.

Hagberg, M. and Jonsson, B. (1975) The amplitude distribution of the myoelectric signal in an ergonomic study of the deltoid muscle. *Ergonomics*, **18**, 311-319.

Hettinger, T. (1985) Occupational hazards associated with diseases of the skeletal system. *Ergonomics*, **28**, 69-75.

Jonsson, B. (1976) Evaluation of the myoelectric signal in long-term vocational electromyography. In: *Komi PV (ed) Biomechanics V-A*, University Park Press, Baltimore, 509-514.

Jonsson, B. (1978) Kinesiology -with special reference to electromyographic kinesiology. *Contemporary Clinical Neurophysiology*, **34(EEG Suppl.)**, 417-428.

Jonsson, B. (1982) Measurement and evaluation of local muscular strain in the shoulder during constrained work. *Journal of Human Ergology*, **11**, 73-88.

Karwowski, W. and Brokaw, N. (1992) Implications of the proposed revisions in a draft of the revised NIOSH lifting guide (1991) for job redesign: a field study. *Proceedings of the Human Factors Society 36th Annual Meeting*, 659-663.

Komi, P.V. (1973) Relationship between muscle tension, EMG, and velocity of contraction under concentric and eccentric work. *in New Developments in Electromyography and Clinical Neurophysiology*, vol. 1, J.E. Desmedt, Ed. (Karger, Basel, Switzerland, 1973), pp. 596-606.

Kumar, S. and Garand, D. (1992) Static and dynamic lifting strength at different reach distances in symmetrical and asymmetrical planes. *Ergonomics*, **35**, 861-880.

Lafortune, D., Norman, R. and McGill, S. (1988) Ensemble averages of linear enveloped EMGs during lifting. *The Fifth Biennial Conference of Canadian Society for Biomechanics*, 92-93.

Lavender, S.A., Tsuang, Y.H., Hafezi, A., Andersson, G.B.J., Chaffin, D.B. and Hughes, R.E. (1992) Coactivation of the trunk muscles during asymmetric loading of the torso. *Human Factors*, **34**, 239-247.

Leskinen, P.J., Stalhammar, H.R., Rautanen, M.T. and Troup, J.D.G. (1992) Biomechanically and electromyographically assessed load on the spine in self-paced and force-paced lifting work. *Ergonomics*, 35, 881-888.

Louhevaara, V., Long, A., Owen, P., Aickin, C. and McPhee, B. (1990) Local muscle and circulatory strain in load lifting, carrying and holding tasks. *International Journal of Industrial Ergonomics*, **6**, 151-162.

Marras, W.S. and Mirka, G.A. (1988) Trunk muscle response to trunk asymmetry, velocity and load level. *Proceedings of the Human Factors Society -32nd Annual Meeting*, 660-664.

Marras, W.S. and Mirka, G.A. (1989) Trunk strength as a function of trunk asymmetry and trunk velocity during concentric and eccentric exertions. *Advances in Industrial Ergonomics Safety I*, 159-156.

Marras, W.S. and Mirka, G.A. (1990) Muscle activities during asymmetric trunk angular accelerations. *Journal of Orthopaedic Research*, **8**, 824-832.

McGill, S.M. (1991a) Kinetic potential of the lumbar trunk musculature about three orthogonal orthopaedic axes in extreme postures. *Spine*, **16**, 809-815.

McGill, S.M. (1991b) Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: implications for lumbar mechanics. *Journal of Orthopaedic Research*, 9, 91-103.

McGill, S.M. and Norman, R.W. (1986) Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine*, **11**, 666-678.

Mirka, G.A. (1991) The quantification of EMG normalization error. *Ergonomics*, **34**, 343-352.

Potvin, J.R., McGill, S.M. and Norman, R.W. (1991) Trunk muscle and lumbar ligament contributions to dynamic lifts with varying degrees of trunk flexion. *Spine*, **16**, 1099-1107.

Putz-Anderson, V. and Waters, T.R. (1991) Revisions in NIOSH guide to manual lifting. Paper presented at national conference entitled "A national strategy for occupational musculoskeletal injury prevention -Implementation issues and research needs." University of Michigan. Ann Arbor, Michigan.

Rosenburg, R. and Seidel, H. (1989) Electromyogrphy of lumbar erector spinae muscles - influence of posture, interelectrode distance, strength, and fatigue. *Eurpean Journal of Applied Physiology*, **59**, 104-114.

Ruhmann, H. and Schmidtke, H. (1989) Human strength: measurements of maximum isometric forces in industry. *Ergonomics*, **32**, 865-879.

Seroussi, R.E. and Pope, M.H. (1987) The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. *Journal of Biomechanics*, **20**, 135-146.

Snook, S.H. and Ciriello, V.M. (1991) The design of manual handling tasks: revised tables of maximum acceptable weights and forces. *Ergonomics*, **34**, 1197-1213.

Sommerich, C.M. and Marras, W.S. (1992) Temporal patterns of trunk muscle activity throughout a dynamic, asymmetric lifting motion. *Human Factors*, **34**, 215-230.

Stalhammar, H.R. and Louhevaara, V. (1992) Anthropometric, muscle strength, and spinal mobility characteristics as predictors in the rating of acceptable loads in parcel sorting. *Ergonomics*, **35**, 1033-1044.

Stokes, I.A.F., Rush, S., Moffroid, M., Johnson, G.B. and Haugh, L.D. (1987) Trunk extensor EMG-torque relationship. *Spine*, **12**, 770-776.

Vink, P., Daanen, H.A.M., Meijst, W.J., and Ligteringen, J. (1992) Decrease in back strength in asymmetric trunk postures. *Ergonomics*, **35**, 405-416.

Vink, P., Van der Velde, E.A. and Verbout, A.J. (1988) A functional subdivision of the lumbar extensor musculature: recruitment patterns and force-RA-EMG relationships under isometric conditions. *Electromyography and Clinical Neurophysiology*, **28**, 517-525.

Waters, T.R., Putz-Anderson, V., Garg, A. and Fine, L.J. (1993) Revised NIOSH equation for the design and evaluation of manual lifting tasks. *Ergonomics*, **36**, 749-776.

Winkle, J. and Bendix, T. (1986) Muscular performance during seated work evaluated by two different EMG methods. *European Journal of Applied Physiology*, **55**, 167-173.

Winter, D.A. (1988) The Biomechanics Motor Control of Human Movement. New York: John Wiley & Sons, Inc.

Yang, J.F. and Winter, D.A. (1984) Electromyographic amplitude normalization methods: improving their sensitivity as diagnostic tools in gait analysis. *Archives of Physical Medicine and Rehabilitation*, **65**, 517-521

APPENDIX A

Review of the Literature

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1.0 Introduction

Occupational lifting has occupied the research efforts of countless authors since it has gained recognition as a threat to the general health of the world's workforce. Due to the building knowledge that past efforts have provided in this field of research, the following literature review attempts to summarize only the most recent, relevant studies. The reader is expected to be knowledgable in the subject of kinesiology and have some understanding of biomechanical principles. The review of literature has been subdivided into the areas of anatomy, biomechanics, electromyography (EMG), EMG signal processing and occupational lifting to provide a comprehensive understanding of the material for the puposes of following the logic of the study's methodology. A strong emphasis has been placed on reporting previous quantitative results.

2.0 Anatomy of the Lumbar Spine

2.1 Basic Structure

To study the EMG response of various lifting postures and/or techniques, a thorough knowledge of the anatomical mapping of the structures involved is vital (Ortengren and Andersson, 1977). There are five sections to the human spine: the cervical, thoracic, lumbar, sacral, and coccyx sections. It is the lumbar section which contains the largest, mobile spinal vertebrae (the sacrum and coccyx being fused). There is a natural lordotic curve in the lumbar region which includes five vertebral segments (Moore, 1985). The vertebral segments of this region contain all the characteristic structures normally associated with spinal vertebra: a spine, transverse processes, superior and inferior facets, pedicle, lamina as well as a vertebral body. These structures contain several points of muscle and ligament attachment which provide the spine with most of its postural stability (Miely *et al.*, 1990). Several muscles appear in the lumbar cross-section including anterior abdominals, lateral obliques and posterior erector spinae incorporating the transversospinalis, interspinalis, intertransversalis, and sacrospinalis muscle groups (Ortengren and Andersson, 1977; McGill and Norman, 1986; Miely *et al.*, 1990). Functionally based lumbar muscle studies concerned with extensor moments have focused on the erector spinae muscle group, termed deep postvertebral muscles (Ortengren and Andersson, 1977), including m. multifidus (Chapman and Troup, 1969; Jonsson, 1970; Andersson *et al.*, 1974; Seroussi and Pope, 1987; Stokes *et al.*, 1987; Vink *et al.*, 1988; Vink *et al.* 1989; Mirka and Marras, 1991; Hamrick and Gallagher, 1992; Leskinen *et al.*, 1992; Stalhammar and Louhevaara, 1992). Reasons for concentrating on the erector spinae include the high incidence of injury in the lumbar region in the occupational setting (Magora, 1974; Brinckmann, 1985; Hettinger, 1985) and that the sacrospinalis specifically acts as the dominant extensor muscle moment generator and thus may be at greatest risk among lumbar all tissues (McGill, 1991a).

2.2 Muscles of the ES

The muscles included in the erector spinae may be separated into a medial, transversospinal muscle compartment containing the multifidi, rotator, interspinal, and intertransversarii muscles, and a lateral, sacrospinalis compartment containing the longissimus and iliocostalis muscles (Ortengren and Andersson, 1977). The m. semispinalis, part of the transversospinal compartment, is poorly developed in the lumbar spine exerting minimal influence on lumbar mechanics and for all practical purposes may be ignored (Ortengren and Andersson, 1977; Miely *et al.*, 1990). Although m. multifidus lies deep to m. semispinalis, it remains prominent in the medial lumbar region and extends the length of the spine. In the lumbar region, m. multifidus originates from the mamilliary processes and inserts two to four levels superior into the spinous processes. The main action for m. multifidus is the maintenance of postural control as well as aiding in extension, lateral flexion, and rotation (Miely *et al.*, 1990). Intermediately, m. longissimus thoracis, the longest of the erector spinae group, arises from the sacrum and transverse processes, inserts into the transverse processes and ribs, and contains two pars. Pars thoracis occurs superiorly; pars lumborum occurs inferiorly and is confluent with fibres of the m. iliocostalis lumborum. Actions of the m. longissimus thoracis pars lumborum include extension and lateral flexion (Miely *et al.*, 1990). The lateral fibres of erector spinae are termed m. iliocostalis lumborum also made up of a pars thoracis and pars lumborum sections. M. iliocostalis lumborum pars lumborum originates on the iliac crest and attaches to the lower six to seven ribs and is responsible for trunk extension and lateral flexion (Meily *et al.*, 1990).

2.3 ES Muscle Function

Muscle function is determined by line of pull between origin and insertion, moment arm or distance the line of pull acts from the joint(s), muscle architecture, cross-sectional area, as well as its contribution to the movement with respect to other muscles crossing the same joint(s). These factors help to determine if a particular muscle is classified as a prime mover, antagonist, synergist, or fixator for any specific movement (Moore, 1985). For lifting tasks, the muscles under investigation act mainly as prime movers but can also be recruited as synergists, fixators, and even antagonists for frontal plane movements (McGill, 1991a). Similar functional characteristics are inherent in the three major bands of the lumbar m. erector spinae yet relative differences may be explained by the different muscles' moment arm lengths over any other factor. Nemeth and Ohlsen (1986) have reported erector spinae moment arm lengths, at the L5/S1 vertebral level, to the bilateral motion axis, about which flexion and extension occurs, of 71 mm, and to the anteroposterior motion axis, about which lateral flexion occurs, of 23 mm for males, using computed tomography (CT) scans. Moment arm lengths were measured from the disk centre to the centre of the muscle's cross-sectional area. Kumar (1988) had slightly different results reporting a 60 mm erector spinae moment arm in the sagittal plane and a 30.3 mm moment arm in the coronal plane for males at the L5 vertebral level. Based on moment arm lengths, erector spinae has a greater mechanical advantage for flexion/extension movements. More detailed work by Tracy *et al.* (1989) provides moment arm information for the three major divisions of erector spinae at the L4/L5 level. In the sagittal plane, anteroposterior distances include: 60.5 mm (n=10) for m. multifidus, 58.3 mm (n=10) for m. longissimus, and 48.1 mm (n=9) for m. iliocostalis. In the coronal plane, lateral distances include: 17.9 mm (n=10) for m. multifidus, 37.5 mm (n=10) for

m. longissimus, and 54.9 mm (n=9) for m. iliocostalis. From this information, it appears that m. multifidus has the greatest mechanical advantage for flexion/extension and m. iliocostalis, the most laterally located erector spinae muscle, has by far the greatest mechanical advantage for lateral flexion movements. Combining EMG information, muscle orientation, cross-sectional area, and moment arm lengths, a better understanding of the functional contributions for the erector spinae components is possible.

An early study by Jonsson (1970) examined the functional differences of the m. multifidus, m. longissimus, and m. iliocostalis at various lumbar levels as well as T12. Jonsson used invasive wire electrodes to record the electromyographic activity of the three columns of muscle at each lumbar spinal level during several different postures. Highlights from this study show the three muscles to have different functions and that sometimes these muscles have different functions at different levels. Some examples include higher activity on the contralateral side, especially from the lateral fibres, during rotation of the trunk and a collective increase in activity during any trunk extensor effort. These conclusions were somewhat tainted by the lack of consistency between subjects regarding the muscle sites tested; sample sizes for specific muscle sites ranged from n=1 to n=10. A study by Andersson *et al.* (1974) revealed no significant functional differences between m. multifidus, m. longissimus and m. iliocostalis at levels L1, L3, and L5 using either surface or indwelling wire electrodes. However, electromyograms were taken during quiet standing or sitting at different levels of inclination and subsequently did not require substantial muscle activation levels. Seroussi and Pope (1987) tested the electromyographic activity between left and right trunk muscle groups in response to moderate moments (no more than 26 N·m external moment) created in the sagittal and frontal planes from static lifting tasks. Collecting bilateral EMG signals at the L3 level, 3 cm from the midline, they found the sum of the right and left m. erector spinae processed EMG signals to be highly correlated ($r^2=0.96$) to the sagittal plane moment and the difference of the right and left processed EMG signals to be highly correlated ($r^2=0.95$) with the frontal plane moment. The data from Seroussi and Pope (1987) suggest a linear relationship between increased moment arm in the sagittal and frontal planes and the symmetrical increase or asymmetrical difference of the m. erector spinae EMG for static lifting tasks.

3.0 Biomechanics of the Spine

3.1 Structural Mechanics of the Spine

The mechanical structure of the vertebral column resembles that of a flexible rod which, in its normal state, contains natural curves that add to its inherent instability (Smith and Fernie, 1991). In order to increase its stability for movements and bearing internal and external loads, the spine is compressed by the contraction of the trunk musculature which connects the pelvis and upper girdle to several points at each vertebral level along the spine's entire length (Marras and Mirka, 1990; McGill, 1990). Ligament forces at range of motion limits (Potvin *et al.*, 1991) and, to a lesser extent, intra-abdominal pressure (Marras and Mirka, 1990) also add to the stability of the spine. Finally, the intervertebral disks themselves contribute to the function of the spine by resisting compression between the vertebral bodies (Chaffin and Andersson, 1991). In lifting tasks, there are several situations which can alter the contributions of these active and passive tissues to the stability of the spine.

3.2 Influence of Posture

Posture has a major influence on how the muscles, ligaments, and intervertebral disks provide support. One aspect of posture is the degree of lumbar flexion. Excellent work by McGill and Norman (1986) showed muscular components to account for 99% of the extensor moment at the L4/L5 level; the disc contributed 1%, while the passive ligamentous tissues provided no restorative moment for lifting tasks where lumbar flexion rarely exceeded 34°. At 59° of lumbar flexion, ligaments and annular fibres collectively contribute up to 89% of the total extensor moment during a 22 kg load lift, while concurrent erector spinae activity occurs at 2% MVC (Potvin *et al*, 1991). In this case, hip extensors posteriorly rotate the pelvis to produce the lifting moment. There is a point where the contributions of passive tissue for the production of the total extensor moment become significant. However, during normal lifting in the sagittal plane where a flat back posture is adopted, the ligamentous system remains disengaged (McGill and Norman, 1986). Another aspect of posture is axial twist. Several studies show decreased maximal back strength in asymmetric trunk postures. Garg and Badger (1986) discovered a 7-22% reduction, with respect to symmetric lifting, of the maximum acceptable loads during three different asymmetric lifting postures. For the same asymmetric postures, maximal isometric strength was reduced 12-31%. More detailed work by Vink *et al.* (1992) included the investigation of 23 separate postures where combinations of trunk flexion, lateroflexion, and rotation were tested against maximal isometric strength measures. Their results showed that the lowest force production occurred during the most extreme measured posture of 30° rotation and 30° lateroflexion with added flexion having no additional influence on the magnitude of the maximum force. Additionally, rectified and averaged EMG values revealed higher activity on the posteriorly displaced muscle origin during trunk rotation as well as higher activity on the upward shoulder side during lateroflexion for maximal extentions. According to Vink *et al.* (1992), these results confirm an interaction between the muscle length, muscle activation and the angular moment of the muscle on trunk extension force production performance. These results do not establish a quantitative relationship but, in essence, rotation and lateroflexion should be avoided during lifting tasks. McGill (1991b) concludes that axial twisting during a lift inhibits the ability to produce maximum muscular contractions due to an increase in co-contraction from muscles not responsible for the torque production. However, erector spinae maintains appreciable potential to generate torque about all three movement axes indicating its mechanical importance for all spinal motions (McGill, 1991a). In fact, erector spinae always produced the greatest amount of activity for asymmetric trunk accelerations from 45° flexion and 30° axial rotation at mean angular accelerations up to 135 deg/s^2 (Marras and Mirka, 1990). The same study showed erector spinae EMG activity, the most active of the lumbar muscles, as high as 50% MVC even though a minimal torque of approximately 4.1 N·m was applied. Also, greater asymmetry lead to increased coactivation from antagonists, and a decrease in magnitude of angular acceleration. Collectively, this information suggests the importance of lumbar spine stabilization over the production of large torque levels for axial twisting. Thus, lifting asymmetrically has its mechanical cost. The manual handling of loads in the median sagittal plane is justified from a mechanical efficiency point of view. The muscles producing an extension moment can be utilized closer to their MVC levels with less passive tissue involvement, when the lumbar spine remains in its natural resting posture which is analagous to upright standing (McGill, 1991b). Emphasizing this point, the shear force in the lumbar spine is significantly higher with a greater degree of lumbar flexion yet remains insensitive to differences between the stoop and squat lifting techniques when the natural lumbar curve is preserved (Potvin *et al.*, 1991).

3.3 Influence of Dynamic Factors

Acceleration of the load is another key component in the lifting task (McGill and Norman, 1985; Bush-Joseph *et al.*, 1988; Gagnon and Smyth, 1992; Tsuang *et al.*, 1992). Tsuang *et al.* (1992) reported a 46% increase of the L5/S1 moment using a dynamic analysis versus a static analysis when lifting a 150 N load at normal speed. Acceleration is simply how quickly the load is moved from its original static position to its maximal velocity during the course of the lift, which is controlled by force application of the musculoskeletal system. Increased acceleration of the load affects all trunk muscles to various degrees. More superficial extensor muscles such as latissimus dorsi are proportionately more active at higher accelerations and the level of coactivation between muscles significantly increases (Marras and Mirka, 1990). Intervertebral disk pressure is also influenced by dynamic lifting factors. Bush-Joseph *et al.* (1988) studied the effects of slow (0.80-0.90 m/s), moderate (1.1-1.2 m/s) and fast (1.7 m/s) lifting speeds, on L5/S1 disc compression while lifting a 150 N load during free lift, leg lift and back lift techniques. They found that disc compression values increased linearly with increased lifting speed. In fact, according to Freivalds *et al.* (1984), acceleration levels seem more influential, than the mass of the load, for increasing compressive forces at the L5/S1 joint. They discovered that heavier loads which were lifted more slowly than lighter loads produced a significantly lower L5/S1 compressive force. This notion is supported by Gagnon and Smyth (1992) who discovered similar increases in joint muscular moments, spinal loadings, and mechanical work with no increased benefits of energy transfers between the load and the musculoskeletal system when harnessing acceleration effects during the lifting task. These studies collectively agree that faster and jerkier lifting motions increase the chance for injury to the lumbar spine due to heightened spinal compressive forces.

4.0 Electromyography (EMG)

4.1 Electrode Placement

Measurement of the myoelectric activity of a muscle requires proper location of the muscle through unique muscle functions requiring a thorough knowledge of functional anatomy. A location method which employs only palpation and inspection can be invalid due to the anatomical variations of an individual (Rozendal and Meijer, 1982). It is also important that the functional test reflect the mechanical context of the investigation, otherwise, the electrode location of a muscle may not remain valid for unrelated movements (Rozendal and Meijer, 1982). The purpose of correct electrode placement is twofold: to achieve minimal crosstalk and to measure as completely as possible the level of neural activity in that muscle (Jonsson, 1978; Vink *et al.*, 1989). Vink *et al.* (1989) concluded that the electrode placement with one pair at level L5, 3 cm lateral to the midline, two pairs at L3, 3 cm and 6 cm lateral to the midline and three pairs at L1, 3 cm, 6 cm and 9 cm lateral to the midline was most effective for the optimal myoelectrical measurement and reduced crosstalk for the medial multifidus, the longissimus, and the lateral iliocostalis intrinsic back muscles. Perhaps this level of redundancy is unnecessary as other authors have represented the lumbar m. erector spinae with only one electrode pair (Seroussi and Pope, 1987) measuring, as well, latissimus dorsi, the oblique muscles and rectus abdominus (Marras and Mirka, 1990; McGill, 1991b; Hamrick and Gallagher, 1992; Lavender *et al.*, 1992). These authors represented the m. erector spinae EMG using recordings that were taken from the L3 or L4 level, 3 cm lateral to the midline. Although incidence of back injury is concentrated at the lumbar vertebral levels, there remains a prevalence of injury occurance to the lower thoracic levels (Hettinger, 1985). Hettinger (1985) researched the incidence of damage to the vertebral column in various occupations and reported that the lower thoracic region was almost as prone to damage as the lumbar region for a variety of occupations. Other authors agree that the measurement of lower thoracic m. erector spinae activity is important to understanding spinal mechanics related to injury. Lafortune *et al.* (1988), McGill (1991a), and Potvin *et al.* (1991) recorded

EMGs from the T9 level, 4 cm lateral to the midline. Lafortune *et al.* (1988) attempted to document the linear envelopes of four trunk muscles and used an electrode placement of 4 cm lateral to T9, 3 cm later to both L3 and L4/L5, and a location over the iliocostalis lumborum pars lumborum lateral to L2. They found similar muscle function for dynamic lifts involving constrained and unconstrained, stoop and squat, and twist and no twist characteristics based on phase comparison of the linear envelopes. There was no investigation of the differences in EMG amplitudes for the four back muscles.

4.2 Force vs. EMG Activity

Electromyography (EMG) offers a way of determining a muscle's involvement for a particular task by measuring its level of activity. To conclude that the level of myoelectric activity measured reflects some biomechanical function, a relationship must be established between EMG values and force output (Jonsson, 1978). It is well known that, in the absence of fatigue, increased force of contraction is related to increased frequency of action potentials of the individual motor units, as well as an increase in the number of motor units involved (Jonsson, 1978). There is continuing debate concerning the relationship between the recruitment of motor units and the muscle's force output (Rozendal and Meijer, 1982). Andersson *et al.* (1977) have shown linear relationships between L3/L4 disc pressure, lumbar EMG amplitudes, and trunk extensor moments within the ranges of 0 to 300 N of applied external load and 0 to 50° of trunk flexion. All three parameters are within limits found in safe occupational lifting. Results from work by

Andersson and Ortengren show that disk pressure, intra-abdominal pressure and myoelectric signals respond similarly and consistently to changes in the trunk extensor moment (cited in Andersson and Ortengren, 1984). Under situations of very high loads and extreme flexion, these variables have shown a curvilinear relationship to one another (Andersson and Ortengren, 1984; Seidel *et al.*, 1987). However, much of this information was derived from data collected during static contractions. A comparison of static to dynamic responses (Garg *et al.*, 1982; McGill and Norman, 1985) have shown dynamic measurements to inflate these variables due to mass accelerations of the load and body segments. It is uncertain exactly how a dynamic response would affect the EMG/force relationship, although it seems the erector spinae maintains a linear relationship whereas latissimus dorsi and rectus abdominous show a very slight curvilinear relationship, especially during large trunk extension accelerations (Marras and Mirka, 1990). For most practical purposes, there is general agreement that a linear EMG/force output relationship exists. Indeed, the EMG/force relationship of the m. erector spinae appears linear up to 90% maximum voluntary contraction (MVC) over a range of trunk flexion up to 40° (Mirka and Marras, 1991).

Thus far, it has been established that m. erector spinae EMG is a feasible indicator of an individual's low back stress during lifting and can be related to safe lifting guidelines based on disc compression values. M. erector spinae EMG could therefore be used as a measure of predicted load on the lumbar spine under specific conditions (Andersson and Ortengren, 1984).

5.0 Amplitude Probability Distribution Function (APDF)

5.1 Rationale of the APDF

EMG must be processed from its raw form by any number of methods to provide information of the force/time history of the muscle's contraction profile (Winter, 1988). There are several methods of signal processing used for representing skeletal muscle electrical activity. An important aspect of any EMG study is the link between the recorded signal and the mechanical performance of the muscle. The basic muscle model contains elements of both amplitude and time of onset, or mechanical delay (Winter, 1976). The amplitude probability distribution function (APDF) is a signal processing technique that, when applied to EMG, describes the distribution of different levels of muscle contraction for the period of EMG recorded (Jonsson, 1982). In terms of the basic muscle model, the EMG amplitude is the only influential element for the APDF. A base level of contraction, normally a MVC, is used to normalize the measured EMG signal to a percent MVC (%MVC) scale. When several contraction samples are taken, the amplitude probability for a certain contraction level is the probability of the EMG being lower than or equal to that contraction level. The APDF has been utilized in ergonomic investigations measuring various muscles in the shoulder and low back (Hagberg and Jonsson, 1975; Jonsson, 1982; Andersson and Ortengren, 1984; Winkle and Bendix, 1986; Louhavaara *et al*, 1990).

5.2 Characteristics of the APDF

The APDF characterizes EMG in the following ways: the lowest contraction level occurs at a probability of 0, whereas the highest contraction level is seen at a probability of 1; more vertical sections of the curve indicate a proportionately large occurrence of the corresponding contraction level; more horizontal sections denote very little occurrence of those contraction levels (Jonsson, 1982). Based on work by Bjorksten and Jonsson (1977), and assumptions regarding muscular endurance, a threshold limit APDF has been suggested for continuous work (Jonsson, 1978). The threshold limit APDF contains three regions for muscle amplitude endurance for prolonged work. Low level ("static") work is defined by probability values close to 0.1 and should not exceed 2-5%MVC; the limit of median force of contraction is defined by probability values close to 0.5 and should not exceed 10-14%MVC; the limit of peak muscular loads are defined by probability values close to 0.9 and should not exceed 50-70%MVC. An APDF could therefore be applied to determine a more specific measure of an individual's capability to perform a lifting task. EMG allows the researcher to obtain a continuous record of the muscle activation profile for a given movement (Jonsson, 1978). In the case of lifting tasks, it is the back extensors which are of main interest (Andersson and Ortengren, 1984; Freivalds *et al.*, 1984; Jorgensen *et al.*, 1985; McGill and Norman, 1985; Louhevaara *et al.*, 1990). There are however, some methodological problems which must be addressed for valid EMG measurement when applied to the APDF.

5.3 Limitations of the APDF

The APDF does not indicate the frequency of muscular contractions. Only the number of observations for each specific contraction level is recorded (Winkle and Bendix, 1986). Thus, identical APDF's could be produced for a continuous static contraction and an intermittent static contraction if the observations are consistently taken during the peak of the intermittent contractions (Winkle and Bendix, 1986). This limitation is prevalent for long-term data collection periods and activities demanding static postural maintenance where the movements are cyclic in nature. It is important, therefore, that the activity recorded be adequately defined in terms of its movements and temporal characteristics.

Muscle fatigue leads to greater EMG activity for the same force output (Hagberg, 1979). This situation is characterized in the APDF by a shift of the profile to higher contraction amplitudes. A fatiguing situation is described by Jonsson (1988) as a prolonged work task which exceeds 5% MVC. Those EMG levels below 5% MVC are termed the static component of work whereas those EMG levels beyond 5% MVC are termed the dynamic component of work. It has been shown that the erector spinae muscle contains a fatiguing static component during lifting tasks (Louhevaara *et al.*, 1990). However, these results were taken under conditions of long-term continuous lifting, thus the APDF of the m. erector spinae was likely tainted with fatigue effects. Methodology that incorporates proper resting periods for prolonged lifting tasks can control for muscle fatigue.

6.0 Aspects of Occupational Lifting

The following subsections address the basis for occupational lifting guidlines, present the more popularly used guidelines, and finally, focus on the development of the

NIOSH guidelines from a biomechanical perspective.

6.1 Source of Occupational Lifting Injuries

Muscle load is the major cause of occupational musculoskeletal injuries (Sejersted and Westgaard, 1988). Several types of muscle load are possible. A prolonged low level static load (Kilbom, 1988; Westgaard, 1988) and a heavy dynamic load (Snook, 1978; Ayoub *et al.*, 1982; Clemmer *et al.*, 1991) represent the two load extremes. In either case, the load can be the result of internal (postural) or external (material handling) forces, or both. There are two basic methods in which the relationship between the load and the musculoskeletal system can be optimized. One is to reduce the load whereas the other is to increase the capacity of the musculoskeletal system (Kilbom, 1988).

Occupational lifting primarily incorporates factors such as load weight, load size, the dimensions of the lift, and lifting frequency (Snook, 1978; Ayoub *et al.*, 1982), all of which may be manipulated to decrease external forces on the worker. Ideally, humans would be well conditioned in strength and flexibility to enhance their capacity for lifting (Burton *et al.*, 1989). However, even conditioned individuals are subject to occupational risks (Kilbom, 1988). Therefore, the parameters of a lift must be maintained within safe, acceptable limits.

6.2 Occupational Lifting Standards

Snook (1978), Snook and Ciriello (1991) and Ayoub *et al.* (1982) have developed two, frequently used lifting standards in industry. Snook has based his lifting guidelines on empirical data obtained from a psychophysics methodology which is concerned with individuals' perceived effort. Basically, these standards were determined by allowing the lifters to define their lifting limits. The underlying assumption is that individuals know their limitations. However, Freivalds *et al.* (1984) determined that subjects are unable to compensate for differences in box size by an appropriate reduction in the load to maintain uniform compressive forces in the vertebral disks. They claim that to rely on the worker's perception of the stresses of lifting a load may not be adequate to protect the person.

6.3 Development of the 1991 NIOSH Equation

NIOSH standards consider biomechanical, physiological and psychological information for the formulation of their lifting equation. Six factors: the horizontal distance from the hands to the midpoint between the ankles (HF); the vertical starting location of the load above the floor (VF); the distance up or down over which the load must be lowered or raised (DF); the degree of asymmetry away from the midsagittal plane (AF); the coupling between the hands and the load (CF); and the frequency of lifts required by the job (FF) make up the components of the 1991 NIOSH equation (Putz-Andersson and Waters, 1991). These components were developed from information originating from the biomechanics, physiology and psychophysics literature which were used to establish criteria that can be applied to the lifting factors in various degrees depending on the lifting circumstances (Waters *et al.*, 1993). The biomechanical criterion is based on 3400N of compressive force at the L5/S1 disc. The physiological criterion is based on a baseline measure of 9.5 kcal/min maximum aerobic capacity for repetitive lifting. Percentages of this baseline value are used depending on the lifting task (Waters *et al.*, 1993). The psychophysical criterion is based on RWL that are acceptable to 75% of the female population. Discussion of the three criteria and their advantages and disadvantages is contained in Ayoub (1992) where the FF affects which criterion limits lifting performance. He states that for compressive forces of 3.34 kN, low frequency lifting of up to 5.5 lifts/min for floor to knuckle height is limiting through biomechanical criterion, after which lifts are limiting through physiological criterion (Ayoub, 1991). Ayoub also includes that psychophysical criterion do not become limiting until 6.37 kN compressive force limit is reached. The biomechanical criterion is also limiting when investigating the HF and AF for low frequency lifting. Kumar and Garand (1992) investigated the effects of reach, symmetry, and body position with respect to lifting strength and found reach, followed by asymmetry, to significantly decrease lifting strength for both men and women. This relationship held true for static and dynamic lifting. In a study by Garg (1989), where the 1981 NIOSH equation was used the HF was also very influential on the MPL values yet did not significantly affect physiological or psychophysical measures for low frequency (0.2 lifts/min) lifting. Due to the inverse relationship between the HF and the MPL values, L5/S1 compression forces remained relatively stable ranging from 3924 N to 4022 N.

7.0 Summary of Literature Review

In its natural posture, the spine can become a very strong mechanism for lifting with proper coactivation of trunk musculature. Lumbar lateroflexion and axial twist can dramatically influence the trunk muscles' ability to produce an extensor moment. The degree of trunk acceleration is crucial for the maintenance of minimal disc compression forces. Linear biomechanical relationships exist between disc compression, restorative moment, and lumbar EMG within moderate loads, accelerations, and lumbar postures which are usually found in lifting tasks. A controlled, technically correct lift is necessary to control for potential sources of EMG measurement error.

EMG presents a viable method for determining the mechanical cost for a lifting task in the absence of passive tissue involvement. It corresponds linearly to normal lifting situations which exist in industrial environments. Electrode placement is an important issue for valid measurement.

The APDF characterizes EMG as the probability of occurrences for each contraction level. Applied to a single dynamic lift, the APDF is not influenced by muscle fatigue or frequency of contractions.

Occupational lifting induces muscle load, a major cause of musculoskeletal injury. Training the individual for lifting is not as effective as altering the factors of the lift. Standards have been developed empirically and objectively to determine the safety limits for lifting. Since individuals are not always good judges of potential risk, objectively derived standards are more appropriate.

Safe lifting guidelines have traditionally focused on disc pressure when biomechanical considerations were included. There is no data concerning the erector spinae muscle attributes. By determining the APDF of the erector spinae EMG, based on the 1991 NIOSH (draft) lifting guidelines, it is possible to conclude which of the normalized amplitudes are within safe lifting limits.

8.0 BIBLIOGRAPHY

Andersson, G.B., Jonsson, B. and Ortengren, R. (1974) Myoelectric activity in individual lumbar erector spinae muscles in sitting. A study with surface and wire electrodes. *Scandinavian Journal of Rehabilitation Medicine Supplement*, **3**, 91-108.

Andersson, G.B.J. and Ortengren, R. (1984) Assessment of back load in assemblyline work using electromyography. *Ergonomics*, **27**, 1157-1168.

Andersson, G.B.J., Ortengren, R. and Nachemson, A. (1977) Intradiskal pressure, intra-abdominal pressure and myoelectric back muscle activity related to posture and loading. *Clinical Orthopaedics and Related Research*, **129**, 156-164.

Ayoub, M.M. (1991) Determining permissible lifting loads: an approach. *Proceedings of the Human Factors Society 35th Annual Meeting*, 825-829.

Ayoub, M.M. (1992) Problems and solutions in manual materials handling: the state of the art. *Ergonomics*, **35**, 713-728.

Ayoub, M.M., Chaffin, D.B., Drury, C.G., Herrin, G.B., Kroemer, A.R., Lind, A.R. and Troup, J.D.G. (1982) Work practices guide for manual lifting. *National Institute for Occupational Safety and Health*, Cincinnati, Ohio.

Bjorksten, M. and Jonsson,B. (1977) Endurance limit of force in long-term intermittent static contractions. *Scandinavian Journal of Work, Environment and Health*, **3**, 23-27.

Brinckmann, P. (1985) Pathology of the vertebral column. *Ergonomics*, **28**, 77-80.

Buckle, P.W., Stubbs, D.A., Randle, I.P.M. and Nicholson, A.S. (1992) Limitations in the application of materials handling guidelines. *Ergonomics*, **35**, 955-964.

Burton A.K., Tillotson K.M. and Troup, J.D.G. (1989) Prediction of low-back trouble frequency in a working population. *Spine*, **14**, 939-946.

Bush-Joseph, C., Schipplein, O., Andersson, G.B.J. and Andriacchi, T.P. (1988) Influence of dynamic factors on the lumbar spine moment in lifting. *Ergonomics*, **31**, 211-216.

Chaffin, D.B. and Andersson, G.B.J. (1991) *Occupational Biomechanics (2nd Edition)*. John Wiley & Sons, Inc, Toronto.

Chapman, A.E. and Troup, J.D.G. (1969) The effect of increased maximal strength on the integrated electrical activity of lumbar erectores spinae. *Electromyography*, **9**, 263-280.

Clemmer, D.I., Mohr, D.L. and Mercer, D.J. (1991) Low-back injuries in a heavy industry I -worker and workplace factors. *Spine*, **16**, 824-830.

Freivalds A., Chaffin, D.B., Garg, A. and Lee, K.S. (1984) A dynamic biomechanical evaluation of lifting maximum acceptable loads. *Journal of Biomechanics*, **17**, 251-262.

Gagnon, M. and Smyth, G. (1992) Biomechanical exploration on dynamic modes of lifting. *Ergonomics*, **35**, 329-345.

Garg, A. (1989) An evaluation of the NIOSH guidelines for manual lifting, with special reference to horizontal distance. *American Industrial Hygene Association Journal*, **50**, 157-164.

Garg, A. and Badger, D. (1986) Maximum acceptable weights and maximum voluntary isometric strengths for asymmetric lifting. *Ergonomics*, **29**, 879-892.

Garg, A., Chaffin, D.B. and Freivalds, A. (1982) Biomechanical stresses from manual load lifting: a static vs. dynamic evaluation. *IIE Transactions*, **14**, 272-281.

Hagberg, M. (1979) The amplitude distribution of surface EMG in static and intermittent static muscular performance. *European Journal of Applied Physiology*, **40**, 265-272.

Hagberg, M. and Jonsson, B. (1975) The amplitude distribution of the myoelectric signal in an ergonomic study of the deltoid muscle. *Ergonomics*, **18**, 311-319.

Hamrick, C.A. and Gallagher, S. (1992) The effects of lifting posture on trunk muscle activity. *Proceedings of the Human Factors Society 36th Annual Meeting*, 742-746.

Hettinger, T. (1985) Occupational hazards associated with diseases of the skeletal system. *Ergonomics*, **28**, 69-75.

Jonsson, B. (1970) The functions of individual muscles in the lumbar part of the spinae muscle. *Electromyography*, **1**, 5-21.

Jonsson, B. (1978) Kinesiology -with special reference to electromyographic kinesiology. *Contemporary Clinical Neurophysiology*, **34(EEG Suppl.)**, 417-428.

Jonsson, B. (1982) Measurement and evaluation of local muscular strain in the shoulder during constrained work. *Journal of Human Ergology*, 11, 73-88.

Jonsson, B. (1988) The static load component in muscle work. *European Journal of Applied Physiology*, 57, 305-310.

Jorgensen, K., Andersen, B., Horst, D., Jensen, S. and Nielsen, A. (1985) The load on the back in different handling operations. *Ergonomics*, **28**, 183-196.

Kilbom, A. (1988) Isometric strength and occupational muscle disorders. *European Journal of Applied Physiology*, 57, 322-326.

Kumar, S. (1988) Moment arms of spinal musculature determined from CT scans. *Clinical Biomechanics*, **4**, 137-144.

Kumar, S. and Garand, D. (1992) Static and dynamic lifting strength at different reach distances in symmetrical and asymmetrical planes. *Ergonomics*, **35**, 861-880.

Lafortune, D., Norman, R. and McGill, S. (1988) Ensemble averages of linear enveloped EMGs during lifting. *The Fifth Biennial Conference of Canadian Society for Biomechanics*, 92-93.

Lavender, S.A., Tsuang, Y.H., Hafezi, A., Andersson, G.B.J., Chaffin, D.B. and Hughes, R.E. (1992) Coactivation of the trunk muscles during asymmetric loading of the torso. *Human Factors*, **34**, 239-247.

Leskinen, P.J., Stalhammar, H.R., Rautanen, M.T. and Troup, J.D.G. (1992) Biomechanically and electromyographically assessed load on the spine in self-paced and force-paced lifting work. *Ergonomics*, **35**, 881-888.

Louhevaara, V., Long, A., Owen, P., Aickin, C. and McPhee, B. (1990) Local muscle and circulatory strain in load lifting, carrying and holding tasks. *International Journal of Industrial Ergonomics*, **6**, 151-162.

Magora, A. (1974) Investigation of the relation between low back pain and occupation. *Scandinavian Journal of Rehabilitation Medicine*, **6**, 81-88.

Marras, W.S. and Mirka, G.A. (1990) Muscle activities during asymmetric trunk angular accelerations. *Journal of Orthopaedic Research*, **8**, 824-832.

McGill, S.M. (1990) Loads on the lumbar spine and associated tissues. In Biomechanics of the Spine: Clinical and Surgical Perspective. Boca Raton, Florida: CRC Press, Inc, 65-95.

McGill, S.M. (1991a) Kinetic potential of the lumbar trunk musculature about three orthogonal orthopaedic axes in extreme postures. *Spine*, **16**, 809-815.

McGill, S.M. (1991b) Electromyographic activity of the abdominal and low back musculature during the generation of isometric and dynamic axial trunk torque: implications for lumbar mechanics. *Journal of Orthopaedic Research*, **9**, 91-103.

McGill, S.M. and Norman, R.W. (1985) Dynamically and statically determined low back moments during lifting. *Journal of Biomechanics*, **18**, 877-885.

McGill, S.M. and Norman, R.W. (1986) Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine*, **11**, 666-678.

Miely, W.R., McLain, R., Weinstein, J.N., Goel, V.K. and Found, E.M.Jr. (1990) Anatomy of the lumbar spine. In Biomechanics of the Spine: Clinical and Surgical Perspectives. Boca Raton, Florida: CRC Press, Inc., 7-35.

Mirka, G.A. and Marras, W.S. (1991) Toward a more accurate discription of the EMG/force relationship of the erector spinae muscle. *Proceedings of the Human factors Society 35th Annual Meeting*, 728-732.

Moore, K.L. (1985) Clinically Oriented Anatomy. Baltimore: Williams & Wilkins.

Nemeth, G. and Ohlsen, H. (1986) Moment arm lengths of trunk muscles to the lumbosacral joint obtained *in vivo* with computed tomography. *Spine*, **11**, 158-160.

Ortengren, R. and Andersson, G.B.J. (1977) Electromyographic studies of trunk muscles, with special reference to the functional anatomy of the lumbar spine. *Spine*, **2**, 44-52.

Potvin, J.R., McGill, S.M. and Norman, R.W. (1991) Trunk muscle and lumbar ligament contributions to dynamic lifts with varying degrees of trunk flexion. *Spine*, **16**, 1099-1107.

Putz-Anderson, V. and Waters, T.R. (1991) Revisions in NIOSH guide to manual lifting. Paper presented at national conference entitled "A national strategy for occupational musculoskeletal injury prevention -Implementation issues and research needs." University of Michigan. Ann Arbor, Michigan.

Rozendal, R.H. and Meijer, O.G. (1982) Human kinesiological electromyography -some methodological problems. *Human Movement Science*, **1**, 7-26.

Seidel, H., Beyer, H. and Brauer, D. (1987) Electromyographic evaluation of back muscle fatigue with repeated sustained contractions of different strengths. *European Journal of Applied Physiology*, **56**, 592-602.

Sejersted, O.M. and Westgaard, R.H. (1988) Occupational muscle pain and injury; scientific challenge (Editorial). *European Journal of Applied Physiology*, **57**, 271-274.

Seroussi, R.E. and Pope, M.H. (1987) The relationship between trunk muscle electromyography and lifting moments in the sagittal and frontal planes. *Journal of Biomechanics*, **20**, 135-146.

Smith, T.J. and Fernie, G.R. (1991) Functional biomechanics of the spine. *Spine*, **16**, 1197-1203.

Snook, S.H. (1978) The design of manual handling tasks. *Ergonomics*, **21**, 963-985.

Snook, S.H. and Ciriello, V.M. (1991) The design of manual handling tasks: revised tables of maximum acceptable weights and forces. *Ergonomics*, **34**, 1197-1213.

Stalhammar, H.R. and Louhevaara, V. (1992) Anthropometric, muscle strength, and spinal mobility characteristics as predictors in the rating of acceptable loads in parcel sorting. *Ergonomics*, **35**, 1033-1044.

Stokes, I.A.F., Rush, S., Moffroid, M., Johnson, G.B. and Haugh, L.D. (1987) Trunk extensor EMG-torque relationship. *Spine*, **12**, 770-776.

Tracy, M.F., Gibson, M.J., Szypryt, E.P., Rutherford, A. and Corlett, E.N. (1989) The geometry of the muscles of the lumbar spine determined by magnetic resonance imaging. *Spine*, **14**, 186-193.

Tsuang, Y.H., Schipplein, O.D., Trafimow, J.H. and Andersson, G.B.J. (1992) Influence of body segment dynamics on loads at the lumbar spine during lifting. *Ergonomics*, **35**, 437-444.

Vink, P., Daanen, H.A.M., Meijst, W.J., and Ligteringen, J. (1992) Decrease in back strength in asymmetric trunk postures. *Ergonomics*, **35**, 405-416.

Vink, P., Daanen, H.A.M. and Verbout, A.J. (1989) Specificity of surface-EMG on the intrinsic lumbar back muscles. *Human Movement Science*, **8**, 67-78.

Vink, P., Van der Velde, E.A. and Verbout, A.J. (1988) A functional subdivision of the lumbar extensor musculature: recruitment patterns and force-RA-EMG relationships under isometric conditions. *Electromyography and Clinical Neurophysiology*, **28**, 517-525.

Waters, T.R., Putz-Anderson, V., Garg, A. and Fine, L.J. (1993) Revised NIOSH equation for the design and evaluation of manual lifting tasks. *Ergonomics*, **36**, 749-776.

Westgaard, R.H. (1988) Measurement and evaluation of postural load in occupational work situations. *European Journal of Applied Physiology*, **57**, 291-304.

Winkle, J. and Bendix, T. (1986) Muscular performance during seated work evaluated by two different EMG methods. *European Journal of Applied Physiology*, **55**, 167-173.

Winter, D.A. (1976) Biomechanical model relating EMG to changing isometric tension. *Digest of the 11th International Conference on Medical and Biological Engineering*.

Winter, D.A. (1988) The Biomechanics Motor Control of Human Movement. New York: John Wiley & Sons, Inc.

APPENDIX B

UNIVERSITÉ D'OTTAWA
UNIVERSITY OF OTTAWA

FACULTÉ DES SCIENCES DE LA SANTÉ
FACULTY OF HEALTH SCIENCES

CONSENT FORM

Comparison of the 1991 NIOSH lifting equation and erector spinae muscle electromyography

This letter is to inform you about the study you are volunteering to participate in, what is expected from yourself as a subject, the potential risks associated with the testing procedure and your rights as a volunteer participant.

The principal investigator of the study is Greg G. Weames (564-9105). The study will be conducted under the supervision of Dr. Peter Stothart (564-5948). This study has been approved by the **Faculty of Health Sciences Human Research Ethics Committee** (Dr. M. Loyer, chair, 451 Smyth Rd., Ottawa, ON, K1H 8M5, 787-6705).

The purpose of this research is to determine if low back muscle activity is within specific limits when measured during safe NIOSH (National Institute for Occupational Safety and Health) lifting tasks calculated from their 1991 lifting equation.

As a volunteer participant of this study you will be asked to wear shorts, tee-shirt and supportive athletic footwear. You will be given a warm-up (stretching) routine and instructions on proper lifting technique. Anatomical measurements (height, weight, knuckle height, age) will be taken. The preparation of the skin (hair shaving, cleaning with electrode gel) prior to the placement of EMG electrodes on the back may cause momentary burning sensations at each of the 6 different sites. You will perform one or more maximal effort deadlifts with an olympic style barbell and weightlifting plates. You will start with 70 kg and step increase 10 kg. You will stop when you decide your maximum lifting weight is reached or when the researcher finds your technique is deteriorating enough to place you at risk. Experimental lifting trials entail 5 different submaximal lifts (not exceeding 23 kg) from floor to waist height, each repeated 8 times in random order for a total of 40 lifts. Frequency of lifting will be set at 1 lift/min.

During the testing procedures you may experience some physical discomfort and fatigue similar to what is experienced during any repetitive lifting task. During the maximal EMG test you will likely experience high levels of physical exertion similar to what is experienced when lifting an immovable object during a maximum effort. You should be aware of the potential risks to some individuals performing lifting tasks such as lightheadedness, fainting, and, very rarely, heart attacks. Lifting objects could subject one to back injury. Every effort will be made by the researcher to ensure proper lifting technique is used throughout the study to minimize the potential injury risks. This includes the use of spotters during the maximal deadlifts. It is your responsibility to inform the researcher of any injury, illness, infection, undue fatigue or other conditions that would prevent you from fully participating in the study. It is also your responsibility to decline outright from participating in this study if you have suffered any back pain or injury within the last year and/or are currently suffering from any chronic back ailments.

All information collected will remain confidential and presented in an anonymous form. You may withdraw from the study at any point without penalty or discrimination. If you feel your rights have been violated please contact Dr. M. Loyer at the above address.

I, ______________________________ (printed name), have read the above comments and wish to proceed as a voluntary subject in the study titled "Validation of the 1991 NIOSH lifting equation utilizing m. erector spinae EMG" taking place at the University of Ottawa, School of Human Kinetics, Biomechanics Laboratory.

NAME______________________________ DATE____________

WITNESS______________________________