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**THE EFFECT OF
WEARING WORK BOOTS
ON
LUMBAR SPINE FLEXION**

by

© Michael Anthony Blench
School of Human Kinetics

**A thesis presented in partial fulfillment of the degree of
Masters of Science, Human Movement Studies,
School of Human Kinetics, University of Ottawa**

May 1998



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DEDICATION

This work is dedicated to my wife, Lucie, for her love, constancy and inspiration and to my daughter, Sandria, for her love, tolerance and understanding.

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Summary

The unilateral muscle activity of the erector spinae muscles, hip extensors, knee extensors, ankle dorsiflexors and plantarflexors, along with joint articulation kinematics of the ankle, knee, hip and lumbar region in the sagittal plane, were examined as a multi-link system. The objective was to determine the effects of wearing work boots on joint kinematics with particular emphasis on the lumbar angle formed by the spinous processes of S3-L3-T10. Seventeen male subjects volunteered to perform specific 'repeated measures' exercises of a material handling nature while wearing properly laced work boots and barefoot with the feet on boot wedges but in an unbound state. Differences in relative angles, and EMG magnitude and timing were examined. The results obtained provide considerable understanding of the more global effects of joint restriction caused by the wearing of necessary workplace apparel. Peak amplitude normalized EMG revealed nothing of significance. Time normalized EMG showed that with the grasping of the container's handles, both the multifidus and biceps femoris had significant differences between wedging and booting, both displaying less muscle recruitment with the wedge. Non-normalized EMG demonstrated high degrees of significance in all muscles except the gastrocnemius, again, with the wedge scenario generally demanding less muscle recruitment. The X,Y-values of the centre of mass were examined and there was significance in the Y-value while no significant changes were obtained in the lumbar angle. There were significant differences in the absolute trunk angle (defined as the angle formed by the spinous processes of T10-S3 and the horizontal) and that of the ankle. Although the lumbar angle did not change, the absolute trunk angle decreased significantly with the wearing of laced work boots. This is a result of compensation for the reduced articulation of the ankle. Reduction of the absolute trunk angle

increases the torque on the spine ³⁷ which could cause cumulative micro trauma for those individuals having to wear work boots as the mainstay.

Relevance

The results of this study demonstrate that by externally restricting the articulation of the ankle - foot complex, other joints of the body compensate. If the restriction is in place for the most part, potentially, derogatory side effects could result such as back problems and the low back could become more susceptible to injury. Any research identifying other etiological factors that could be contributing to back pain or injury is very relevant.

Keywords: Lumbar spine; Manual material handling; Footwear; Biomechanics; low back injuries; EMG signals.

Introduction

Lifting and spinal loading have been studied extensively. These studies have generally concentrated on factors such as load, horizontal distance of the load from the body, trunk extension velocity, pelvic tilt, angle of lumbar lordosis, moment arms, compressive forces, modeling of the spine, metabolic energy components, and electromyography (EMG). In the literature, lowering has received notably less attention. Neither lifting nor lowering of loads have been associated with the wearing of work place apparel or equipment. Nor have they been studied with joint restriction as the independent variable. The long term effects of repeated lifting / lowering strategy on the spinal column or on the globality of the human body has also not been given much attention. However, associations have been formulated between types of occupations and low back pain / injury. Interestingly, associations between the occupations, their demands of apparel / equipment and low back pain / injury have not been formulated.

Only a few lifting / lowering studies¹³ have concerned themselves with muscle phasing and intersegmental co-ordination. Using intersegmental kinematics and electromyography, the research in this article demonstrated that externally imposed articulation restrictions of the most distal intersegmental joint, the ankle, can have significant effects on more proximal segments when performing a manual material handling exercise.

Extensive qualitative observation in several large physiotherapy clinics involved with the rehabilitation of injured workers noted that there appeared to be a correlation between back injuries and the prolonged wearing of work safety boots⁷ (defined as having steel toe caps, shank and heel counters). Examination of captured statistical data revealed a Phi-coefficient between the two dichotomous variables of the wearing of safety work boots and back problems as:

$$\phi = 0.44$$

This would indicate that there exists evidence that there is a correlation for people wearing work boots for long periods to have back problems. Darlington ¹¹ states that "....small and moderate correlations are usually more important than is often realized". May, Masson and Hunter ³⁵ suggested that exploratory research between seemingly unrelated variables yielding correlations in the 0.40's, is high.

These findings are fairly consistent with those of Riihimaki, Tola, Videman and Hanninen ⁴⁰ whose research showed that machine operators / drivers and construction carpenters were the most prevalent in reporting low back pain. Shannon ⁴² summarized the physical or objective findings by stating: "Studies consistently show that low back pain is more common in heavy manual workers than in others. As a result, construction workers have a higher incidence...." Recently published statistics from the Ontario Workers' Compensation Board (OWCB) ⁴³ showed that 56% of all injuries covered by OWCB involve the back. Occupationally, heavy manual workers (including some health care workers) account for nearly 71% of all injuries in the workplace during this period. Both Riihimaki et al. ⁴⁰ and Magora ^{32, 33} pointed out that occupational factors - both objective and subjective - and the wide degree of variability amongst these factors as being of importance in the etiology of low back pain. Shannon ⁴² also discussed the "psycho-social" factors and the perceptions of workers.

General observations performed by qualified professionals have noted that ankle articulation, both dorsiflexion and plantarflexion, are considerably restricted with the wearing this type of boot. Magee ³¹ observed identical restriction with the ankle in a close packed position. However, the manner and extent to which the boot is laced is also an influencing factor. Also, the protective

steel shank of the boot effects the flexing of the sole such that it functions as three segments (Magee ³¹ divides the foot into three sections - hindfoot, midfoot and forefoot) - toe box, midsection, and heel counter - joined by two 'hinges'. The first 'hinge' between the heel and the midsection, has virtually no flexing capability due to the thickness of material of the heel and the proximity of the steel insert in the sole. This hinge lies approximately under the anterior end of the transverse tarsal joint which demonstrates the inversion / eversion movements but virtually no dorsiflexion / plantarflexion movements ⁴. The hinge between the midsection and toe box lies approximately under the metatarsophalangeal joints which demonstrates extension and flexion capabilities. In effect, once the foot is placed in the average work boot and it is laced to about midpoint, the foot is in a device which only permits flexing at one point - the 'hinge' between the toe box and the midsection. All other significant foot joints for sagittal plane movement are held in a state of virtual immobility. Supportingly, Rowe ⁴¹ within his discussion of ankle motion, and Kendall, McCreary and Provance ²³ specifically comment on the importance of the rigidity of the sole. This implies that the articulation of the ankle becomes of paramount importance as it is the next most proximal joint

In contrast to the above, Wichmann ⁵⁰ reports that the billion dollar sports shoe industry designs more flexibility into the front of the shoe (toe box area) to enhance the push-off ability of the runner; walking shoes are more firm in this area so that the wearer rolls off the toes and joints as opposed to bending through them. Midsole cushioning and heel counter (the socket into which the heel fits) design / fit are also of extreme importance. Heel wedges of 15 mm, minimum, are considered necessary for the protection of the Achilles tendon. The traditional work boot has no flexibility in the toe box area; midsole cushioning and arch support is a non-function of the

protective steel shank; the heel counter is also made of steel. Wichmann⁵⁰ quotes the Athlete's Footwear Test Center's recommendations on four attributes for footwear: cushioning, stability, wearability and fit.

Marr and Quine³⁴ in a study of the problems of wearing safety footwear (footwear that only included steel toe caps), reported that 91% of the wearers had one or more foot problems and that foot problems increased with length of wear. Excessive heat (65%) of all respondents, inflexible soles (52%), shoe weight (48%) and pressure from the steel toe cap (47%) were the major findings.

The excessive heat buildup results in a high prevalence to skin breakdown and subsequent skin related problems. The inflexible soles, the shallowness of the steel toe cap were attributed to the lack of bending of the foot for the performance of specific tasks - ladder climbing, fast movement, crouching, bending, crawling and remaining in confined spaces - were specifically singled out. The weight of the footwear was reported to be attributed to the complaints of aching legs. As reported by Ebbeling, Hamill and Crussemeyer¹⁴ and alluded to by Wichmann and Martin⁵⁰, the kinematics of the lower extremity body segments change as well as other physiological factors with changes in the flexing of the foot and heel height.

The Israeli military, with its concern of 'overuse' injuries amongst infantry recruits, initiated a study¹⁷ on the appropriateness of their footwear. Their findings were a 17.6% incidence of overuse foot injuries for those recruits wearing basketball shoes versus 34% for those wearing infantry boots. They concluded that "people with high lower extremity demands may experience a greater degree of transient foot swelling. This along with increased shearing or other local forces,

coupled with improperly fitted shoe wear, may contribute to a higher incidence of overuse foot injuries."

None of these studies correlated or associated 'foot problems' with other bodily injuries or complaints. However, the United States Army Combat Developments Command, 1964 ⁴⁶, and the Medical Research Council in their Industrial Fatigue Research Board of 1926 ³⁶, reported that improper clothing could cause marked increased muscular activity and alter the centre of gravity. Their study showed a higher heel position shifted the centre of gravity forward and caused muscles to work harder against the pull of gravity when in flexion.

Several investigators ^{9, 16, 19, 20, 25, 26, 52} noted biomechanical lumbosacral functional changes with such factors as age, gender, degeneration and previous injury. A comprehensive review of spinal movement in White and Panjabi ⁴⁹ illustrated that the L5-S1 joint is the primary joint for flexion / extension and L4-L5 is the next most mobile. The literature, however, demonstrates a considerable range of findings (a function of measurement technique and method) with regard to range of motion (ROM) for specific spinal joints. Yet consistency lies in the ranking of the ROM by joint. Therefore, any cause of functional changes would appear more significantly in these joints.

Clinically, it has been found that full plantarflexion is reduced in a high percentage of cases where the wearing of work boots is the norm. It appears that the tibialis posterior, which assists in plantarflexion and inversion, has gradually become weaker for concentric contractions, allowing the foot to increase pronation and throw the entire body out of alignment ³⁹

The human body presents a lever system designed around a compromise for speed, co-ordination and range of motion. With this design in mind, it is not difficult to imagine that the

lifting and lowering functions increase the body's risk to injury, whether from compressive forces or tensile forces on soft tissue. Much of today's manual work demands fast, repetitive lifts which could, with mechanical disturbances and increased speed, cause serious muscular disadvantage, resulting in cumulative micro-trauma and increase fatiguability, and again, facilitating the potential for injury.

Biering-Sorensen ⁵ stated that poor endurance of back extensor muscles is a good predictor of future low back pain. However, if forced to work in a mechanically restricted fashion, these muscles will be jeopardized even more thereby increasing the reoccurrences of injury patterns.

Perrott ³⁸, stated that there are practical ways to avoid unnecessary fatigue and trauma . His first suggestion was to eliminate unnecessary movement. "Repetition of even minute movements can be damaging". He recommended to position the body so prime movers, synergists, fixators and antagonists may each be used in the proper function (and timing).

With regard to these findings, there is a potential to modify the biomechanics at this level with natural and / or external forces. In this case, the wearing of standard safety work boots could be one of these external forces. Therefore, consideration must be given to any cause, such as equipment or apparel, that alters this natural flow of events, thereby increasing human body risk.

This study was designed to determine the kinematic changes occurring in a multilink model, that result from the most distal joint's (ankle) articulation has on it an externally imposed restriction (work boots). Also, are there significant changes in muscle recruitment when mechanical restrictions are imposed?

Methods and Materials

Subjects

Seventeen males with no previous history of injuries to the ankles, knees, hips and back during the previous two years, volunteered to participate in this study. They were asked to read and sign the 'Letter of Information and Consent' form (Appendix B). Their age ranged from 19 through to 54 years. Male subjects were used in this study as this gender wears safety work boots in greater numbers than females, at this time. As wide an age range as possible was deemed appropriate to dispel any skewing of results due to age.

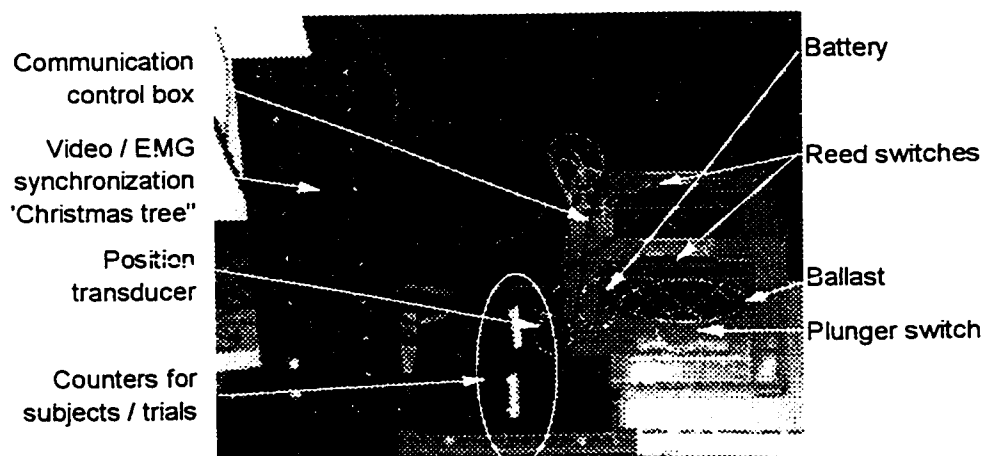
The subjects' age, height and body weight were recorded. Range of motion (ROM) of the ankle, knee, hip and lumbar spine were also taken in accordance with the methodologies of the American Academy of Orthopaedic Surgeons, 1965. Table 1 presents these findings and also gives the average pertinent ROM as determined by the American Academy of Orthopaedic Surgeons for comparison. Circumferences were measured at the height of the symphysis pubis, umbilicus and chest. Anthropometric lengths of the foot, shin, thigh, trunk, upper arm, forearm and hand were measured on the dominant side. All measurements were done in accordance to the Anthropometric Standardization Reference Manual (1988)³⁰. Appendix 'C' summarizes this information.

Table 1. Summary description of subjects' measures and ranges of motion (ROM)

Measurement	Mean	SD	Range	AMA Guides
Age (yrs)	35.6	9.7	19-54	n/a
Height (cm)	181.0	6.3	173.0-194.7	n/a
Weight (kg)	79.8	14.7	59.3-112.5	n/a
ROM: Ankles (deg)	78.2	9.0	61.7-96.0	66.0
ROM: Knees (deg) (flexion)	123.2	11.5	109.3-156.0	134.0
ROM: Hips (deg) (flexion)	108.9	13.8	71.0-126.0	113.0
ROM: Lumbar Spine (deg)	61.9	10.5	39.0-79.3	50.0

Equipment and Materials

A milk bag, plastic, bulk carrying container (Figure 1) with outside dimensions 33 (L) x 33 (W) x 28 (H) cm and weight of 1.7 kg was used. Four handles are built into the mold for ease of handling and it is designed for stacking.

**Figure 1:** Instrumented container

Two, serially connected, reed switch mechanisms were placed on the handles of the container. Their purpose was to initiate a visual and electronic time reference when both handles were grasped. A plunger switch mechanism was placed on the underside of the container to time mark the loss and re-establishment of the container's contact with the floor. A positional transducer from CELESCO Transducer Products Inc., Canoga Park, CA, USA, model: PT101-0050-1111-1110 (sensitivity: 19.092 mV/V/inch) was used to detect the start of the container's upward motion, the apex of the motion and the start and the cessation of the downward motion.

A foot placement grid (Figure 2) constructed of lines drawn 2.5 cm square was used to have the subject's feet in approximately the same position between exercise sessions. This was done by tracing the foot's outline with a wax pencil. The surface was covered with a clear sheet of Plexiglas for abrasion protection and ease of cleaning.

Sufficient sand bagged ballast was placed in the bottom of the container to bring the total weight to 11.4 kg (≈ 111 N). This is well within the 1962 ILO (International Labour Organization) maximum limits for men (143 N) as well as being below the 'optimum condition' lift of the 1981 NIOSH 'action limit' equation (392 N)¹⁰

A Sony Video8 CCD-F501 camera on a tripod was used to record the exercise sessions (see Figure 2, for setup) and a Nikon F-601, 35 mm camera with F: 3.5, 28-200 mm zoom lens was used to record discrete events as appropriate.

A Compaq 386 computer with Promatek's VISION 3000™ software, which uses the University of Michigan's 2-D Static Strength Prediction Model, and a 'PRESENTER[®]' frame-grabber board installed, was used to examine the video tape. This computer was also

equipped with the University of Ottawa's BIOAD software and a Techmar PGL40 analogue-digital signal converting board for the capturing of the electromyography.

A 486DX2®, IBM® personal computer was used to process the EMG signals using the University of Ottawa's BIOPROC software.

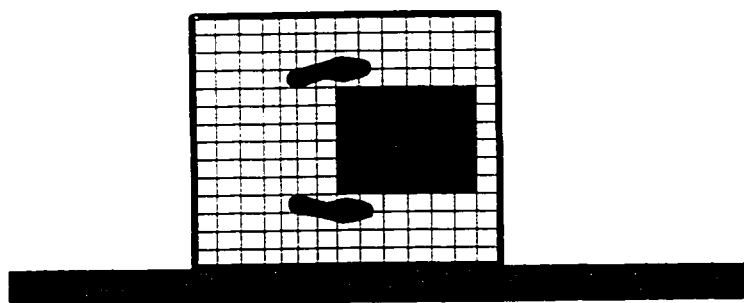
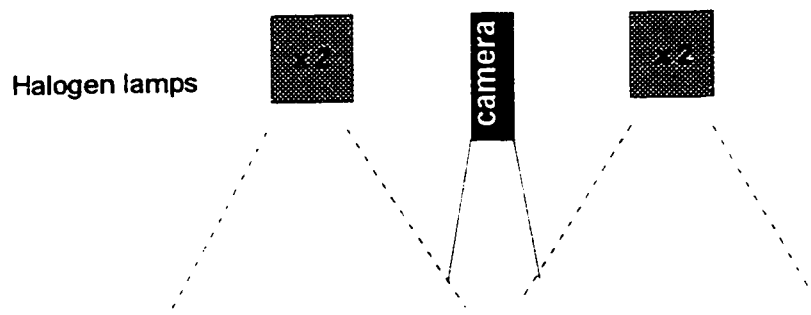
Two metre sticks mounted - one horizontally, one vertically - provided reference points for size and position for each video frame. Body markers of 3M® reflective tape was used to mark the toe, metatarsophalangeal joint, heel, malleolus and the joint centres of the knee, hip, shoulder, elbow and wrist as well as the anterior and posterior superior iliac spine^{8, 47}. 'WELSH' electrodes, with the bulb painted reflective white were used to mark the spinal reference points of S3, L3 and T10 for determining spinal flexion. AquaSonic Gel™ was used to help maintain the seal for these electrodes.

Six pairs of disposable EMG surface electrodes (Ag - AgCl, Blue Sensor) connected to six bioamplifiers (MEGA 3000) were used to collect the muscle activity of the longissimus thoracis (LT), multifidus (MLT), vastus medialis (VM), biceps femoris (BF) (long head), tibialis anterior (TA) and the gastrocnemius (GAS) (lateral head). Each pair of electrodes was positioned 2.5 cm apart, centre-to-centre, aside of the specific muscles' motor points^{12, 48}. EMG sampling rate was at 600 Hz per channel based on the findings of Lamontagne and Coulombe²⁸. Video taping was done at 30 frames per second; shutter speed of 1/1000th second.

Due to the number of muscles monitored and the limitations of the available equipment, bilateral EMG capture although desired, was not possible, therefore, symmetry of muscle recruitment was assumed. Two grounding electrodes were positioned: one, midpoint of the tibia; second, spinous process of T10. This double grounding arrangement was found to require less

CAMERA POSITION

NOTE: Height from the floor of the camera lens axis is $\frac{1}{2}$ subject's height



Back drop with horizontal and vertical metre sticks as reference.

Figure 2: Elevated perspective of the basic hardware positioning for the filming sessions.

gain being applied to the more distal electrode pairs as well as giving assurance that the grounding was maintained during this dynamic exercise.

Anthropometric measures were taken using a leveled platform scale, a moveable anthropometre, a sliding caliper anthropometre, a spreading caliper anthropometre and an anthropometric tape measure. A Cybex EDI 320 (Electronic Digital Inclinometre) was used for the determination of gross range of motion (ROM) measurements.

Experiment

Once the electromyography surface electrodes were in place and tested, the subjects were then asked to approach the container and repetitively grasp, raise, lower and release it ten times, each, during two separate sessions. Ten minute rest periods separated each session. 'Freestyle', but symmetric, squat lifts / lowers (i.e., no postural or mechanical tutoring was given prior nor during the exercise) were used during each session. Lifts / lowers where the heel did not stay on the floor were discarded from the results. These were rare occurrences (4). Each lift / lower cycle comprised seven phases: (1) the preparatory phase where the subject starts the downward motion to grasp the handles; (2) the time interval between the grasping of the container's handles and the start of the container's departure from the floor; (3) the time interval between the start of the raise and the moment the subject was erect (upward movement of the container ceased); (4) time interval where the container is at the apex; (5) the time between the start of the lowering to the time the container touched the floor; (6) the time interval from container touching the floor to the release of the containers handles; (7) post amble where the subject returns to the vertical. Figure 3 highlights the two EMG windows of study as they relate to the phases and events of the exercise.

One session was done barefooted (baseline for comparisons) with a heel wedge and sole identical to those of the participants workboots; the second was done with the boots laced as the manufacturer intended them to be laced. The foot position on the grid of the initial session was marked using a wax pencil so as to maintain approximate alignment intersessionnally. The subjects were requested to perform the sessions in random order so as to eliminate or minimize systematic errors.

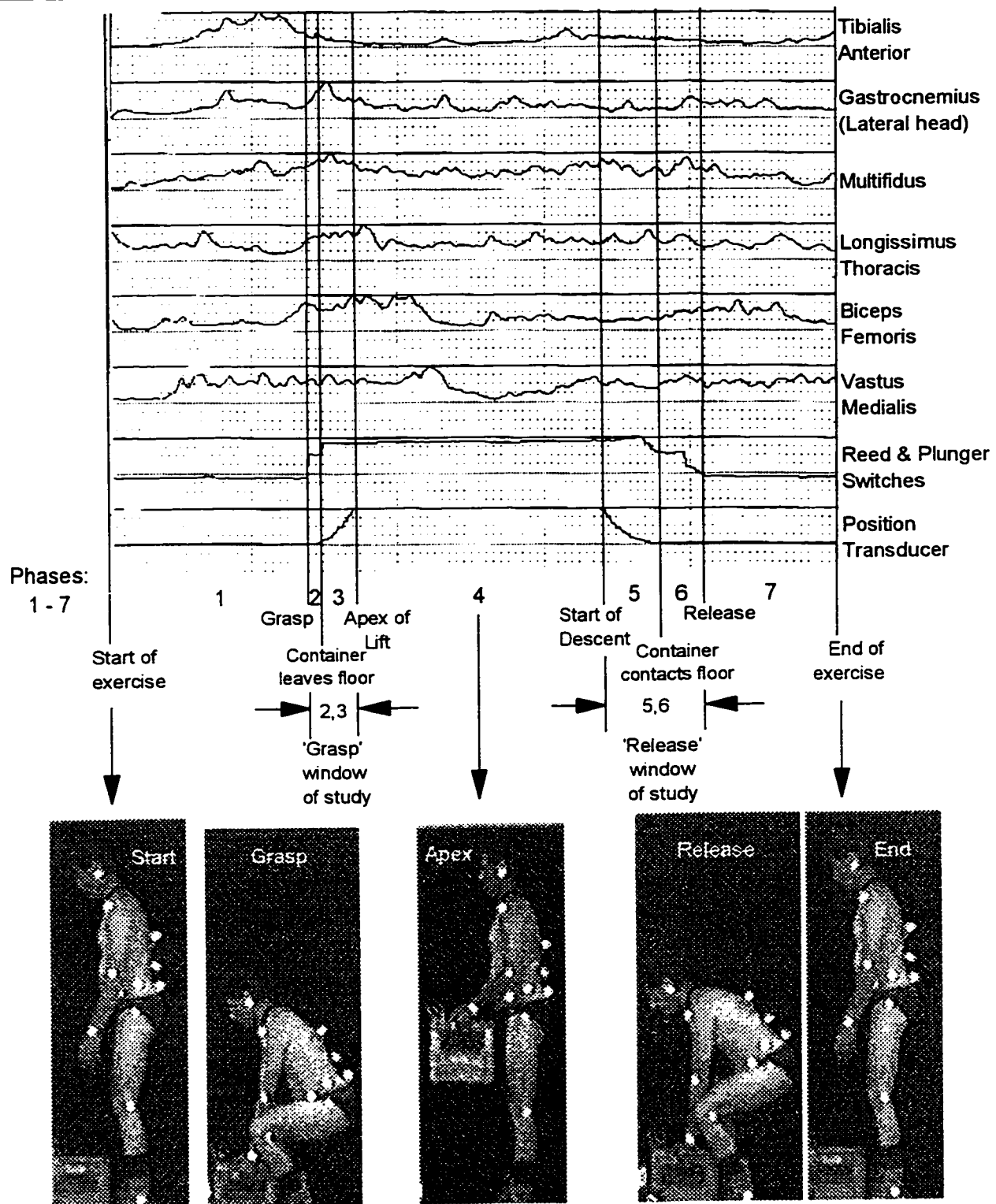


Figure 3: Diagram showing the windows of study in relation to the EMG phases and events of the exercise.

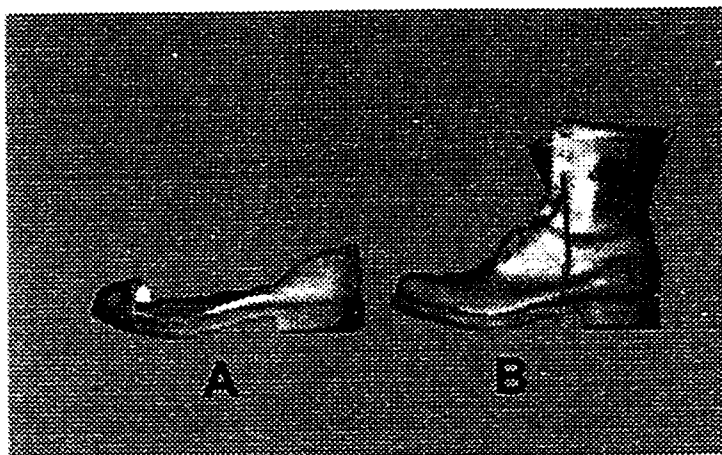


Figure 4: The footwear used in the experiment. The 'wedge' (A)
and the 'boot', (B)

Data Reduction

Electromyography

The raw EMG signals were recorded throughout the prescribed motion and saved on the hard disk drive of the computer and later processed by 'BIOPROC' software. From this continuous EMG signal of the entire exercise cycle, two 'windows' were extracted for analysis. The first was the 'lift' (the grasping (G) of the container's handles through to the cessation of the upward motion or apex of the lift. The second was the 'lower' (the commencement of the downward motion of the container through to the releasing (R) of the handles). These raw EMG signals were A/C high-pass filtered (5 Hz), bias removed using a pure moving average, full wave rectified, and low-pass filtered (5 Hz, zero-phase lag, critically dampened filter, dual pass, 4th order) to produce a linear envelope (LE). The ensemble averages of the LE-EMG of each subject

were integrated (Simpson's) (ILE-EMG) for the computation and comparison of the areas under the curves.

Normalization of the ensemble averages was performed as follows: (1) 100% time normalized; (2) peak amplitude normalized; (3) no normalization at all (i.e. real-time). Global means and other descriptive statistics were gathered.

Video and Anthropometry

The VISION 3000™ system provided a means of computing distances and angles directly from the video images. The angles are computed as per the methodologies of the American Academy of Orthopaedic Surgeons ³ (see Appendix D). These features were used to capture inter-body segment angles and distances at various stages of the lift / lower cycles; specifically the 'grasping' (or 'lift') of the container and its 'release' (or 'lower') after the lowering for kinematic analysis. Standard descriptive statistics were obtained from the anthropometric measures ³⁰ performed as part of the protocol.

The 'Trunk Absolute Angle', (frontal plane, S3/T10 segment) was computed using the technique of Chaffin et al. and Gilad ^{10, 18} which relates all angles to the horizontal plane.

Statistical Analysis

Using the GB-STAT™ statistical software from Dynamic Microsystems, Maryland, USA, repeated measures ANOVA with one factor ('Wedge' versus 'Boot') was performed on the following discrete parameters:

- ♦ Lumbar angle (S3-L3-T10).
- ♦ Trunk absolute angle (frontal plane, S3-T10 segment in relation to Earth's horizontal plane).
- ♦ Centre of mass of the trunk (T10): X-value, Y-value.
- ♦ Ankle, knee and hip angles.
- ♦ EMG: not normalized (i.e. 'real' time) on all muscles monitored.
- ♦ EMG: 100% time normalized on all muscles monitored.
- ♦ EMG: peak amplitude normalized on all muscles monitored.

All of the above angular (degrees) and positional (centimetres) parameters' related ANOVA's were performed at the 'grasp' moment as well as the 'release' moment. EMG ANOVA's were performed on the lifting and lowering windows.

From the repeated measures ANOVA's, the probabilities have been accompanied with Eta² (η^2), an index of the strength of the relationship between the independent variable and the dependent variable.

A Type I error of $\alpha = 0.05$ was selected as the threshold for significance for each ANOVA comparison. Choosing $\alpha = 0.05$ provides a compromise between the stringency of Type I errors while reasonably minimizing Type II errors (β). In the author's opinion, Type II errors are more important in the early stages of exploratory research such as in this manuscript. As the accumulated knowledge increases, the emphasis towards Type I errors should occur.

Results

Relative Lumbar Angle

The lumbar angle by definition, is formed by the spinous processes of S3 - L3 - T10, as viewed at right angles to the sagittal plane. There were no significant differences between the wedged versus booted scenarios while doing the grasp, nor were there significant differences during the release. Table 2, highlights the results.

Table 2. Lumbar angle summary (degrees)

ANOVA	W-Mean	B-Mean	Probability	Eta ²
WG-BG	173.20	172.40	0.16	0.9
WR-BR	173.3	173.3	0.91	0.8

Legend:

W - Wedge G - Grasp * - Significance ($\alpha \leq 0.05$)
 B - Boot R - Release

Absolute Trunk Angle

The absolute trunk angle is by definition the angle formed by the spinous processes of S3 - T10 and the horizontal as viewed at right angles to the sagittal plane. The wedged versus booted scenario while doing the grasp was significant ($p=0.05$). The release wedge versus booted relationship is also very significant ($p=0.03$) as summarized in Table 3.

Table 3. Absolute trunk angle summary (degrees)

ANOVA	W-Mean	B-Mean	Probability	Eta ²
WG-BG	48.5	42.7	0.05*	0.8
WR-BR	43.1	39.2	0.03*	0.9

Legend:

W - Wedge G - Grasp * - Significance ($\alpha \leq 0.05$)
 B - Boot R - Release

Ankle, Knee and Hip Angles

Table 4, shows that the ankle, wedge versus boot while grasping is significant, ($p=0.04$). The ankle is also significant ($p=0.05$) on the release. There were no significant differences in the knees and hips.

Table 4. Ankle, knee and hip angles summary (degrees)

ANOVA	W - Mean	B - Mean	Probability	Eta ²
WAG-BAG	29.3	26.0	0.04*	0.8
WKG-BKG	86.7	82.4	0.34	0.8
WHG-BHG	112.8	113.7	0.32	0.9
WAR-BAR	27.0	24.0	0.05*	0.9
WKR-BKR	79.7	75.4	0.32	0.8
WHR-BHR	110.6	111.7	0.22	0.9

Legend:

W - wedge A - ankle G - grasp * - Significance ($\alpha \leq 0.05$)
 B - boot K - knee R - release
 H - hip

Centre of Mass of the Trunk

The Centre of Mass of the trunk in this study is defined as the T10 spinous process. Its movement in the X,Y-axes was considered and is presented in Tables 5A and 5B representing the grasp and release, respectively. Similar, significant results ($p=0.01$) occur in the Y-axis only.

Table 5A. Grasp: Centre of Mass (T10) summary

ANOVA	W-Mean (cm)	B-Mean (cm)	Probability	Eta ²
WG-BG (X)	80.6	80.0	0.40	1.0
WG-BG (Y)	-108.9	-106.1	0.01*	0.9

Legend: * - Significance ($\alpha \leq 0.05$)

W - Wedge R - Release X - X-coordinate
B - Boot G - Grasp Y - Y-coordinate

Table 5B. Release: Centre of Mass (T10) summary

ANOVA	W-Mean (cm)	B-Mean (cm)	Probability	Eta ²
WR-BR (X)	78.7	77.6	0.20	1.0
WR-BR (Y)	-105.6	-102.2	0.01*	0.9

Legend: * - Significance ($\alpha \leq 0.05$)

W - Wedge R - Release X - X-coordinate
B - Boot G - Grasp Y - Y-coordinate

Electromyography

Table 6 presents the results of time normalized, peak amplitude normalized and real time ANOVA ILE-EMG comparisons for the individual muscles.

The peak amplitude normalized results are unremarkable with no significant differences being noted. Time normalized results reveal that the MLT, in the grasping scenario of wedge versus boot, is significant ($p=0.05$). Also, the biceps femoris is significant ($p=0.01$) in the same scenario. Real time (i.e., not normalized) results highlight many significant differences with the TA, MLT, LT and BF all displaying significantly less muscle recruitment with the wedge while grasping. And, the MLT, BF and VM being significantly different during the release with the MLT and VM recruiting more muscle with the wedge and the BF, less (Table 7).

Table 6. ILE - EMG Summary of repeated measures ANOVA

ANOVA Condition	100% Time Normalized	Peak Amplitude Normalized	Not Normalized (Real Time)
	Prob.	Prob.	Prob.
TA/WG-TA/BG	0.20	0.61	<0.001***
TA/WR-TA/BR	0.77	0.76	0.68
GAS/WG-GAS/BG	0.27	0.32	0.32
GAS/WR-GAS/BR	0.23	0.74	0.28
MLT/WG-MLT/BG	0.05*	0.62	<0.001***
MLT/WR-MLT/BR	0.36	0.24	<0.001***
LT/WG-LT/BG	0.87	0.25	<0.001***
LT/WR-LT/BR	0.20	0.34	0.78
BF/WG-BF/BG	0.01**	0.79	<0.0001***
BF/WR-BF/BR	0.15	0.44	<0.0001***
VM/WG-VM/BG	0.10	0.22	0.61
VM/WR-VM/BR	0.09	0.27	<0.0001***
Eta ² ≥ 0.86		Eta ² ≥ 0.46	Eta ² ≥ 0.66

Legend: TA - Tibialis Anterior
 GAS - Gastrocnemius
 MLT - Multifidus
 LT - Longissimus Thoracis
 BF - Biceps Femoris
 VM - Vastus Medialis

W - Wedge
 B - Boots
 G - Grasp
 R - Release

* p≤0.05; ** p≤0.01; *** p≤0.001

Table 7. Significant Differences Summary of the Real Time ILE-EMG

Grasp		Release	
Muscle	Activity	Muscle	Activity
TA (p<0.001)	W < B		
MLT (p<0.001)	W < B	MLT (p<0.001)	W > B
LT (p<0.001)	W < B		
BF (p<0.0001)	W < B	BF (p<0.0001)	W < B
		VM (p<0.0001)	W > B

Legend: TA - Tibialis Anterior

W - Wedge

GAS - Gastrocnemius

B - Boot

MLT - Multifidus

LT - Longissimus Thoracis

BF - Biceps Femoris

VM - Vastus Medialis

Discussion

In the present study, seventeen male subjects were studied on a repeated measures, material handling activity to determine the differences, if any, in the lumbar angle, absolute trunk angle, angles of the ankle, knee and hip, centre of mass spatial positioning and the EMG activity of six muscles while grasping and releasing the object.

Angles:

As anticipated, the results show that when the ankle is bound by the laced boot, there are significant differences in the angle in which the ankle is capable of flexing (grasping: $p=0.04$ and releasing: $p=0.05$). Although these changes in the ankles' performance do not significantly alter the angles of the knees, hips and lumbar angle, they do, very significantly, alter the spatial positioning of the centre of mass (CMT) of the trunk (defined as the T10 spinous process) in the Y-axis (both grasp and release: $p=0.01$) and in the absolute trunk angle (ATA) ($p=0.05$, grasping; $p=0.03$, releasing). Figures 5 and 6 demonstrate the differences in the Y-axis (height) and the absolute trunk angle (ATA). The horizontal lines drawn through the body markers show the differences in the height and the lines drawn through the body markers attached to S3 and T10 show the change in the ATA. The 'booted' posture consistently presents a shallower ATA; the CMT is in a higher position as is the S3 body marker. These are a direct consequence of the reduced ankle articulation causing the body to compensate which is contrary to what Perrot³⁸ recommends. Because the ankles flex more with the wedge, the trunk stays in a more upright position and with the buttocks lower in relation to the floor. With the added articulation restriction of the boot, the ankle is not flexing as much. The findings demonstrated that the knee and hip angles are approximately the same between the two scenarios of wedge and boot. This

means that the buttocks have to shift posteriorly and are not as close to the floor. To compensate for this weight shift, the trunk advances anteriorly (pivot point is the ankle as all other angles are the same) to maintain the body's balance and to permit the arms to reach the object (when it is in contact with the floor).

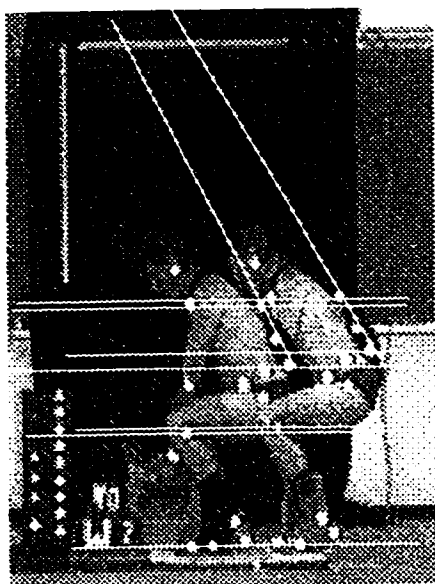


Figure 5:

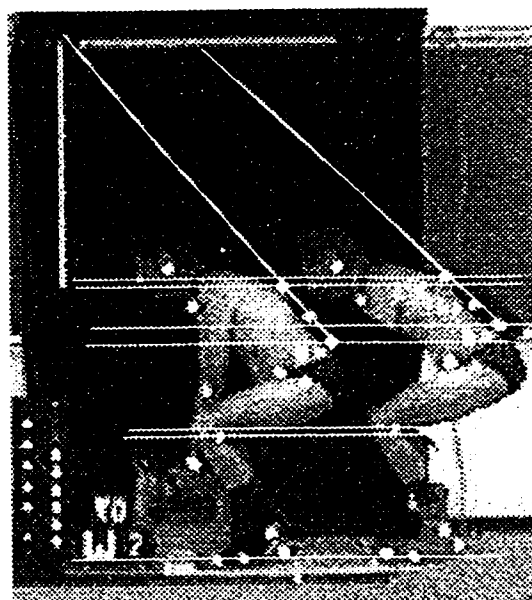


Figure 6:

Superimposed frames of 'grasping' (Figure 5) and 'releasing' (Figure 6) of the wedge (leftmost image) and the boot (rightmost image,) respectively. The horizontal pairs of lines, from top to bottom, show the vertical displacements (wedge vs boot) of the T10 (CM), S3 and knee body markers. The acute lines in each of the Figures demonstrate the differences in the ATA.

Herdman ²¹ pointed out that for the body's normal spatial alignment and balance, three elements are in equilibrium - bilateral vestibular sensory information, visual and proprioceptive sensory information. In this experiment, changed visual aspects (wedge body position Vs boot body position) and changed proprioception changes (boot Vs wedge) accounted for the changed spatial alignment. Herdman ²¹ elaborated on the importance of the ankle flexion and the quality of

the support surface to the corresponding body position. The quality of the support surface greatly influences the proprioceptive information. This influences the type and quality of muscular recruitment.

The lumbar angle, defined as that formed by the S3 - L3 - T10 spinous processes, did not reveal any significant differences between the wedge and the boot (Table 2). This is due to the consistency of manner in which all of the subjects performed the squatting / semi-squatting (lordotic lumbar spine) as opposed to stooping (flexing of the lumbar spine). As Table 2 shows, the difference in grasping between the wedge and the boot is 0.8° ($173.2^{\circ} - 172.4^{\circ}$). For the release, there was no measurable difference. Research shows that once the trunk or ATA is inclined to the 'critical point'⁴⁵ of 45° ^{29,44} forward, the posterior ligamentous system becomes taut and unloads the erector spinae muscles. The mean ATA during the wedge-grasp is greater than this 45° which means the thoracolumbar musculature is active; with the boot-grasp it is less than 45° and resting on the ligamentous system. Twomey and Taylor⁴⁵ discuss the bowstring effect of the activity of the MLT, LT and the iliocostalis. They also discuss the 'brace' effect (antiflexion) of the transverse abdominis and the antiflexion functionality, 'hydraulic amplifier mechanism', of the thoracolumbar fascia as a result of lumbar back muscle contraction. Twomey and Taylor maintain that when large weights are not being lifted, as is the case in this experiment, the back muscles are capable of controlling flexion.

Real-life presents many compounding factors that would augment the laboratory findings. People have many anthropometric^{22,51} and morphological differences²¹. Ages²¹ vary as do states of physical conditioning and the presence (or absence) of other pathologies. The wearing of other restrictive clothing (i.e. tightly fitting jeans) or equipment (i.e. a tool pouch) would effect the

results as would the performing of the exercise on unlevel or uneven ground ²¹. Other variables that would effect the results are the asymmetry ², speed ²¹, and precision demands of the movement, the physical size of the object being handled, its actual (and perceived) weight, type of handles, etc.

Electromyography

Peak amplitude normalization, for this set of circumstances, does not reveal anything profound. The smallest repeated measures ANOVA probability was $p=0.22$ for the VM when grasping. What is of interest, though, is that the η^2 index is still moderately high, (Table 6) which means that more than 46% of the variation in the ILE-EMG can be accounted for by the effect of the independent variable of the wedge - boot..

For the 100% time normalized (TN), ILE-EMG presents not only significance with the MLT and the BF while grasping ($p=0.05$ and $p=0.01$, respectively), but also demonstrates the reason why the time element, which has a very widespread variance during the grasping and releasing windows, has to be eliminated for rational and consistent comparisons. Table 7 summarizes the findings of the not time normalized or 'real-time' (RT) ILE-EMG. The TA, MLT, LT and BF all are significantly different while grasping between the wedge and the boot. As in the TN grasping findings, those that are significant demonstrate less muscle recruitment with the wedge than with the boot. The MLT and BF have common significance between the TN and RT. This is also present in the RT release scenario, however, the MLT has greater muscle recruitment with the wedge (the opposite of the grasping).

Examination of the actual average times to perform the tasks of grasping and releasing revealed essentially no differences between the wedge and the boot (on average, ≤ 0.005 s difference). The intra and inter subject consistency in timing performance was astounding and unanticipated. This means that the tasks are 'naturally' time normalized. This 'natural' normalization also explains why the peak amplitude normalized results were so unremarkable. With both time normalization and the peak amplitude normalization occurring at the same time, one can expect very unremarkable results.

Eccentric muscle contractions as in the releasing, are known to demand less muscle recruitment than concentric contractions ¹⁵ (as in the grasping). Table 8 summarizes the respective comparative mean muscle recruitments between the grasping and the releasing. The experimental findings supported this notion of 'eccentric less than concentric'. *Prima facie* evidence of the TA seems to contradict this notion, however, in the grasping motion, the TA is in fact eccentrically contracting.

Table 8: 100% Time Normalized ILE-EMG - Comparative, Mean Muscle Recruitments Between Grasping and Releasing

Muscles	Grasping (G)	Recruitment <, =, >	Releasing (R)
Tibialis Anterior	G	<	R
Gastrocnemius	G	>	R
Multifidus	G	>	R
Longissimus Thoracis	G	>	R
Biceps Femoris	G	>	R
Vastus Medialis	G	>	R

Legend: < less than; = equal to; > greater than

Individual muscle recruitment comparisons between the wedge and the boot are summarized in Table 9. These muscle recruitments appear to be somewhat haphazard in

Table 9: 100% Time Normalized ILE-EMG - Comparative, Mean Muscle Recruitments Between the Wedge and the Boot

Muscles	Wedge (W)	Recruitment <, =, >	Boot (B)
Tibialis Anterior	W	<	B
Gastrocnemius	W	>	B
Multifidus	W	<*	B
Longissimus Thoracis	W	>	B
Biceps Femoris	W	<*	B
Vastus Medialis	W	>	B

Legend: < less than; = equal to; > greater than

* - Significance($\alpha \leq 0.05$)

comparison to the orderliness of the grasp / release (Table 8). However, upon examination, the patterns are perfectly logical when one considers the following:

- ♦ The MLT will be recruited more with the boot due to the more horizontal, and consequently more unstable, attitude of the trunk: $W < B$.
- ♦ The BF with the boot is in a more eccentric mode due to the trunk's position. It has to recruit more to perform as expected: $W < B$.

Examination of the 'real-time' EMG revealed some interesting information. Several muscles demonstrated significance (reference Table 6 for details). Table 10 shows that the recruitment was less for the grasp in all cases due to the fact that the duration of the grasp was a

fraction - three tenths - of the duration of the release. In real terms, this translates to a greater fatigue factor ^{15, 24} with the release even though the amplitude of the recruitment may be less (eccentric contractions).

Table 10: Not Normalized ILE-EMG - Comparative, Mean Muscle Recruitments Between Grasping and Releasing

Muscles	Grasping (G)	Recruitment <, =, >	Releasing (R)
Tibialis Anterior	G	<	R
Gastrocnemius	G	<	R
Multifidus	G	<	R
Longissimus Thoracis	G	<	R
Biceps Femoris	G	<	R
Vastus Medialis	G	<	R

Legend: < less than; = equal to; > greater than

Examining the muscle recruitment between the wedge and boot is also very interesting in that similar trends still exist albeit for possibly more subtle reasons. Table 11 summarizes the findings.

Table 11: Not Normalized ILE-EMG - Comparative, Mean Muscle Recruitments Between Wedge and Boot

Muscles	Grasp: Wedge Vs Boot	Release: Wedge Vs Boot
Tibialis Anterior	<	=
Gastrocnemius	<	>
Multifidus	<	>
Longissimus Thoracis	<	=
Biceps Femoris	<	<
Vastus Medialis	<	>

Legend:

< less than

= equal to

> greater than

From Table 11, 75 % of the cases show that the wedge requires less than or equal to the required recruitment of the boot. The intricacies of the muscles of this non-normalized information would require more investigation, however, it is suspected that GAS is recruiting more with the wedge release due to a greater antagonistic role with increased dorsiflexion. The VM performed a similar role to the GAS in that its antagonistic role amplified with the wedge release.

The MLT is to working harder for reduced lateral stability at the ankle being amplified by the distance between the ankle and the lumbar spine. If the instability were more proximal, the MLT would be working less. Herdman ²¹ in her section on movement strategies, discusses compensatory musculature firing patterns in relation to postural perturbations.

The stiffness of the boots' material and manner of construction helps to store and release energy to the lower extremity. The storing (when being restricted by the boot) works against the individual while the release is advantageous.

Sources of Error

Surface EMG attenuates the signal because of adipose tissue, the natural conductivity of the subject's tissue ¹, and the placement of the electrodes. With a dynamic movement, EMG can also be subject to spurious artifacts ⁵¹ caused by the movement of electrical leads, and skin movement under the electrodes resulting in improved or lessened electrical contact. Every step possible was taken by the experimenter to control these possible sources of error.

The positioning of body markers could introduce error. Skin movement during the exercise could also be a source of error. With the manual digitization process, the determination of the centroids of the body markers could bring in error. The VISION 3000™ software has been validated with a $\pm 1.5^\circ$ error factor ⁶.

These possible sources of error could work in a compounding fashion or in some form of cancellation. Every effort was made to control them.

Conclusions

The results of this study, involving seventeen male subjects, demonstrated that the wearing of laced work boots profoundly alters the performance of the human body. Although the lumbar angle did not change significantly, its orientation, a function of the absolute trunk angle, did change. This postural change coupled with elements such as age, fatigue, rapid pace and long term repetitiveness³⁷, could cause cumulative micro-trauma²² which results in the low back being more susceptible to injury²⁷.

This awareness should lead to the development of preventative measures to reduce the potential for injury, i.e.:

- ♦ improving the articulation of this type of work boot (and footwear in general) with a new design and / or the improved use of materials.
- ♦ a re-examination of work and everyday living practices, i.e.: the wearing of a safety shoe if the boot is not an absolute necessity; do not wear other restrictive apparel such as tight jeans.
- ♦ a re-examination of the design of required tools and equipment to minimize asymmetrical loading on the body.
- ♦ prompting a 'wellness' concept - morphological changes, physical conditioning and flexibility.

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Appendix A

INTRODUCTION

After extensive, qualitative observation in a clinical setting, there appears to be a correlation between back injuries and the prolonged wearing of work boots (steel toe-capped, healed and shanked) as used in the construction industry as well as other heavy industry (similar observations have been independently made concerning the prolonged wearing of cowboy boots). This clinical observation was quantified by the examination of 1396 patient files at a Ottawa based Community Clinic accredited by the Ontario Workers' Compensation Board. The results of this examination are revealed in Table 1:

Table 1: Statistics from an Community Clinic accredited by the Ontario Workers' Compensation Board.

Number of patients	Remarks
1396	- total number of patients examined.
734	- number of patients with back problems in the lumbar or thoracic area.
662	- patients with other kinds of musculo-skeletal problems.
242 (33%)	- patients with back problems wearing safety work boots. 80% of these patients are workers in the construction trades (skilled or unskilled) or truck drivers (including heavy equipment drivers). The remaining 20% were composed of machine operators, machinists or mechanics.
240 of the 242 patients were of male gender; mean age = 38.8 years; minimum age: 18; maximum age: 62.	

The Phi-coefficient between the two dichotomous variables of the wearing of safety work boots and back problems at this particular clinic was: $\phi = 0.44$.

This would indicate that there exists more than a slight tendency for people wearing work boots for long periods to have back problems. Darlington (1990) states that "....small and moderate correlations are usually more important than is often realized". May et al (1990) suggest that exploratory research between seemingly unrelated variables yielding correlations in the 0.40's, are high.

These findings are fairly consistent with those of Riihimaki et al (1989) whose research showed that machine operators / drivers and construction carpenters were the most prevalent in reporting low back pain. Shannon (1992), summarized the physical or objective findings by stating: "Studies consistently show that low back pain is more common in heavy manual workers than in others. As a result, construction workers have a higher incidence, as do nurses and nursing aids who must often lift patients....". Recently published statistics (1992) from the Ontario Workers' Compensation Board (OWCB) showed that 56% of all injuries covered by OWCB involve the back. Occupationally, heavy manual workers (including some health care workers) account for nearly 71% of all injuries in the workplace during this period. Both Riihimaki (1989) and Magora (1972, 1973) pointed out that occupational factors - both objective and subjective - and the wide degree of variability amongst these factors as being of importance in the etiology of low back pain. Shannon (1992) also discussed the "psycho-social" factors and the perceptions of workers.

General observations have noted that ankle articulation, both dorsiflexion and plantarflexion, are considerably restricted with the wearing of this type of boot. However, the manner and extent to which the boot is laced is also an influencing factor. Also, the protective steel shank of the boot effects the flexing of the sole such that it functions like three segments - toe box, midsection, and heel - joined by two 'hinges'. The first 'hinge' between the heel and the midsection, has virtually no flexing capability due to the thickness of material of the heel and the proximity of the steel insert in the sole. This hinge lies approximately under the anterior end of the transverse tarsal joint which demonstrates the inversion / eversion movements (Basmajian, 1982), but virtually no dorsiflexion / plantarflexion movements. The hinge between the midsection and toe box lies approximately under the metatarsal phalangeal joints which demonstrates extension and flexion capabilities. In effect, once the foot is placed in the average work boot and it is laced to about midpoint, the foot is in a device which only permits flexing at one point - the 'hinge' between the toe box and the midsection. All other significant foot joints for sagittal plane movement are held in a state of virtual immobility. Supportingly, Rowe, (1985), in his discussion of ankle motion, specifically comments on the "...rigid sole of the average working shoe."

In contrast to the above, Wichmann (1993) reports that the billion dollar sports shoe industry designs more flexibility into the front of the shoe (toe box area) to enhance the push off ability of the runner; walking shoes are more firm in this area so that the wearer rolls off the toes and joints as opposed to bending

through them. Midsole cushioning and heel counter (the socket into which the heel fits) design / fit are also of extreme importance. Heel wedges of 15 mm, minimum, are considered necessary for the protection of the Achilles tendon. The traditional work boot has no flexibility in the toe box area; midsole cushioning and arch support is a non-function of the protective steel shank; the heel counter is also made of steel. Wichmann quotes the Athlete's Footwear Test Center's recommendations on four attributes for footwear: cushioning, stability, wearability and fit. From discussions with several retailers of work boots, two criteria appear to surface as being of importance to the purchaser: fit (i.e. does the foot go into the boot) and price (the lower the better).

The work place environment, which calls for the wearing of this type of footwear, often provides more opportunities for the worker to actually work in more extreme body attitudes or positions, as well as doing prolonged periods of physically abusive body movements which are often highly repetitive. Under these circumstances, it would be logical to assume that optimal performance requires movement in as free and as natural a manner as possible, unrestricted and unencumbered by artificial means.

The wearing of any weighty, protective apparel or equipment, causes the human body to work harder. It can also cause restricted articulation and potentially abnormal muscle activity and / or usage. All of these could, over time, manifest into a derogatory situation. Analogously, Ebbeling et al (1994), in a study to determine the effects of varying heights of high-heeled shoes on lower extremity

mechanics and energy cost while walking, found that ankle plantar flexion, knee flexion, vertical ground reaction force and the maximum anteroposterior braking force increased with heel height. They also discovered that heart rate and oxygen consumption increased with heel height.

Thus, it is speculated, that ankle articulation restrictions as well as reduced sole flexing are, in fact, causing the worker to do 'normal' movements in a subtly abnormal fashion leading to postural fatigue, rendering the back liable to cumulative trauma.

Statement of the Problem:

The purpose of this research is to determine quantitatively, the relationship between ankle and foot joint restriction and its effect on range of motion (ROM) of the knee, hip and lumbar joints - particularly, L5-S1 and L4-L5 - as well as electromyography (EMG) of erector spinae (longissimus thoracis, multifidus), vastus lateralis and medialis, biceps femoris (long head), tibialis anterior and gastrocnemius (lateral head) muscles during controlled, symmetrical, manual material handling.

Justification for the Study:

According to the Ontario Workers' Compensation Institute (1992), back pain is the leading cause of chronic disability for people under 45 years of age and the third most common cause between the ages of 45 and 64. In Ontario, in 1990, back pain accounted for 50% of all WCB claims. Low back injuries for the same period in the United States account for an estimated cost of \$11 - 18 billion.

25%-28% of work injuries involving lost time compensation claims are back injuries - \$575 million for back injuries alone in 1987 (Ontario Workers' Compensation Board, Backfacts, 1988). Each injury costs approximately 40-45 sick days per year on the average (Backfacts, 1988). It is felt that any research identifying any other etiological factor(s) and presenting a possibility of reducing or preventing back pain or injuries, would be a worthwhile endeavour, morally, socially and economically.

Potentially, the results of this study could be instrumental in redesigning work boots so as to be complimentary to the preceding paragraph.

Increased knowledge of the effect of apparel on joint restriction also justifies this study.

REVIEW OF RELATED LITERATURE

Statistics of Low Back Pain:

The astonishing proliferation of low back pain (LBP) in industrialized societies is probably the major public health problem of the century (75-80% of population). Literature as early as the year 1900 document findings, theory and discussion covering a multitude of aspects of LBP ranging from psychosomatic causes, psychology, training programmes, anatomical causes, surgical procedures, medical examination techniques, therapy programmes, etc. Gracovetsky and Farfan (1986), quoting other sources stated: "...the most common disability in persons under the age of 45; in those over 45, it is third after arthritis and heart disease most back problems are work-related". This is confirmed in the introductory letter of John Frank, Director of Research, Occupational Back Pain: Epidemiological Perspectives, Ontario Workers' Compensation Board.

The statistics on the economic effects of low back pain and spinal injury are somewhat vague, and staggering. This perhaps is due to the associated intangible or indirect costs.

The most recent statistics encountered are from the Ontario Workers' Compensation Board 1993 Annual Report Supplement, which are summarized in Table 2.

Table 2: Summary of published statistics from the Ontario Workers' Compensation Board 1993 Annual Report Supplement

Claims registered in 1993:		368,485 (100%)
Number Allowing lost time:		125,122 (35%)
Age group contrib. to <u>greatest</u> no. of claims:		30-34 (17.1%)
Gender breakdown:		70.4% male 29.5% female
<u>Greatest</u> number of lost time claims by body part:		34.0% back (incl. neck)
Claim <u>greatest</u> cause:		32.6% overexertion
<u>Greatest</u> source of injury	18.5% - 'bodily motion'. - Gracovetsky (1986), says that in surgical series, 65% are "torsional injuries" and 35% are "compression injuries".	
<u>Greatest</u> contributors by occupation:		- service** 15.3% - fabricating** 13.9% - transport** 8.5% - clerical 7.6% - machining** 6.8% - constr'n** 6.6% - MMH** 6.3% - med /health* <u>6.1%</u> Total: <u>71.1%</u>
* denotes that some wear protective footwear (PF).		
** denotes the wearing of PF as a rule.		
Avg. no. of days lost during life of an injury:		84.6 days
No. of days lost due to back related claims:		~ 3.6 million mandays

Lost man days are only a small part of the economics. The cost of property and equipment damage, litigation, down time, loss of customers and goodwill, third party damages and other hidden costs, result in massive increases in the numbers by orders-of-magnitude. As a result, the incentives for determining definitive, causative or associated etiological factors that can be controlled to reduce the numbers, are very strong.

Marr and Quine (1993) in a study of the problems of wearing safety footwear (footwear that only included steel toe caps), reported that 91% of the wearers had one or more foot problems and that foot problems increased with length of wear. Excessive heat (65%) of all respondents, inflexible soles (52%), weight (48%) and pressure from the steel toe cap (47%) were the major findings.

The excessive heat buildup results in a high prevalence to skin breakdown and subsequent skin related problems. The inflexible soles, the shallowness of the steel toe cap were attributed to the lack of bending of the foot for the performance of specific tasks - ladder climbing, fast movement, crouching, bending, crawling and remaining in confined spaces - were specifically singled out. The weight of the footwear was reported to be attributed to the complaints of aching legs. As reported by Ebbeling et al (1994) and alluded to by Wichmann et al (1993), the kinematics of the lower extremity body segments change as well as other physiological factors with changes in the flexing of the foot and heel height.

The Israeli military , with its concern of 'overuse' injuries amongst infantry recruits, initiated a study on the appropriateness of their footwear (Finestone et al,

1992). Their findings were a 17.6% incidence of overuse injuries for those recruits wearing basketball shoes versus 34% for those wearing infantry boots. They concluded that "people with high lower extremity demands may experience a greater degree of transient foot swelling. This along with increased shearing or other local forces, coupled with improperly fitted shoe wear, may contribute to a higher incidence of overuse foot injuries."

None of these studies correlated nor associated 'foot problems' with other bodily injuries or complaints. However, the United States Army Combat Developments Command, 1964, and the Medical Research Council in their Industrial Fatigue Research Board of 1926, reported that improper clothing could cause marked increased muscular activity and alter the centre of gravity. Their study showed a higher heel position shifted the centre of gravity forward and caused muscles to work harder against the pull of gravity when in flexion.

Anatomy:

Movement in a safe and supported manner is the key function of muscles. For this to occur, some muscles act as primary movement producers while others are recruited, timed / phased / sequenced to ensure the movement is performed in conformance to this key function.

For spinal extension (and lateral bending) in this study, the prime 'torque producers' (Jull et al, 1992; Basmajian et al, 1985; Warwick et al, 1973) that are being monitored are the longissimus thoracis. The longissimus thoracis (LT) and the spinalis thoracis (ST), which is intimately blended to the LT, are the prime extensors of the thoracic and lumbar regions. Only recently have the origins and attachments of the LT been reported correctly in the literature. Lee, (1989) described that "these two muscles are not confined to the thoracic spine, but span several segments to gain attachment to the posterior superior iliac spine of the innominate bone and to the aponeurosis of the erector spinae muscle which covers the lumbar longissimus muscle before attaching to the posteroinferior aspect of the sacrum. Consequently, these muscles are also capable of influencing the motion of bones which do not directly articulate."

The stabilizing function of the lumbar spine is accomplished primarily by the multifidi (Gracovetsky et al, 1977, 1985; Macintosh et al, 1987; Tess et al, 1987; Zetterberg et al 1987; Warwick et al, 1973; Jull et al, 1992; Basmajian et al, 1985).

The gluteus muscle group and the hamstrings are well documented as being hip extensors (Basmajian et al, 1985; Warwick et al, 1973)

Sullivan, (1989), reported that the " thoracolumbar fascia (TLF) provides a major support mechanism for lifting, regardless of the lumbar posture adopted. It is known that because of the TLF's attachment to muscles and its role as a ligament, it has "the best mechanical advantage of all lumbar tissues that provide anti-flexion movement." Added tension along the course of the TLF is facilitated by intra-abdominal pressure.

Although the quadriceps femoris is the group of prime knee extensors, only two of the muscles - vastus medialis (VM) and vastus lateralis (VL) - are included in this study. The EMG action of the quadriceps as well as that of the gluteus muscles and hamstrings during lifting, is well understood (Németh et al, 1984; Basmajian et al, 1985; Speakman et al, 1977; Trafimow et al, 1993; De Looze et al, 1993), but the action in relation to the gastrocnemius (GN) and tibialis anterior (TA) has not been studied.

The gastrocnemius and the tibialis anterior - prime plantarflexor and dorsiflexor / invertor, respectively (Warwick et al, 1973; Basmajian et al, 1985) - have hardly been examined, to date, with regard to any complex, bodily movement.

Normal, average ranges of motion (ROM) are depicted in the following Table 3 (American Academy of Orthopaedic Surgeons, 1965). The average ranges cannot accurately be determined due to the wide variation in the degrees of motion amongst individuals of varying morphologies and statures and age. Therefore, Table 3 serves only as a guide.

Table 3: Average range of joint motion in degrees of arc.

Joint	Avg.
Ankle:	
Plantarflexion	48
Dorsiflexion	18
Knee:	
Flexion	134
Hyperextension	10
Hip:	
Flexion	113
Extension	28
Spine - Thoracic & Lumbar:	
Flexion	85
Extension	30

There are a variety of flexion / extension / rotation / measurement techniques. Undoubtedly, X-ray is the most accurate, however, it is invasive which increases patient risk and it is costly. Other techniques are either surface contact or remote and involve digitized film / video, mechanical and electric goniometres / inclinometres, and 'skin stretch' by steel tape measuring. Studies done by Lamontagne et al., 1988, demonstrate that electrogoniometry techniques show no significant difference from the results obtained by X-ray measurement and therefore, are the best available technology for this type of study considering factors such as cost, time, and freedom of mobility.

In the literature, it can be reviewed that there exists many studies of the lumbar spine with specific regard to osteokinematic and arthrokinematic function (Bogduk, 1980, 1986, 1987; Lovett, 1903; Meadows, 1985; Farfan, 1978).

Farfan (1973), Burkart (1979), Yasuma (1990), Gilmore (1986), Grieve (1988), Kirkaldy-Willis (1979, 1983), Panjabi (1978) and others noted biomechanical lumbosacral functional changes with such factors as age, degeneration and previous injury. A comprehensive review of spinal movement in Panjabi and White, (1990), illustrates that the L5-S1 joint is the primary joint for flexion / extension and L4-L5 is the next most mobile. The literature, however, did demonstrate a considerable range of findings (a function of measurement technique and method) with regard to range of motion (ROM) for specific spinal joints. Yet consistency lies in the ranking of the ROM by joint. Therefore, any cause of functional changes would appear more significantly in these joints.

Clinically, it has been found that full plantarflexion is reduced in a high percentage of cases where the wearing of work boots is the norm. It appeared that the tibialis posterior, which assists in plantarflexion and inversion, has gradually become weaker for eccentric contractions, allowing the foot to increase pronation and throw the entire body out of alignment (Rasch et al, 1971). It is easy to postulate such a consequence will cause knee and / or low back problems.

The literature review has also revealed another important factor influencing the spinal mechanics. Burton and Tillotson, (1988), reported that even in healthy adult subjects, sagittal flexion mobility range in males of 34-54 years of age is less than

that of females. Women retain extension range into middle age and demonstrate a steady degradation of ROM. Men, in this study, were seen to have lost upper lumbar mobility to a greater degree. Lower lumbar mobility was found to decline for both genders in middle age. It is of related interest, that Lee (1989), referring to the works of other authors, reports that degeneration at the sacroiliac joint begins as early as the fourth decade of life for men and the fifth decade for women.

Mechanics:

Sullivan, (1989), reported that after reviewing existing literature on lifting, a flexed posture (posterior tilt of the pelvis) protects the erector spinae and multifidus muscles from excess stress. He also stated, in the presence of a herniated intervertebral disc, a lift in flexion increases tensile forces on the posterior annulus, while a lift in extension (anterior pelvic tilt) increases the intradiscal pressure.

It is the premise, that with decreased ankle dorsiflexion due to the physical restriction of joint mobility by the wearing of work boots, the lumbar spine is not able to achieve its normal flexion range for a safe and repetitive lift. That would allow the use of the ligamentous system thereby protecting the erector spinae and multifidus of excess stress. Morris, Lucas, Bresler, (1961), Troup, (1977), and Gracovetsky, (1988), all noted that loads greater than 68 kg (667 N) required flexion of the lumbar spine in the initial phase of the lift. The initial, crucial phase being that effort to overcome inertia. At this time, ground reaction forces are also greatest, according to Frievolds, Caffin, Grag et al. (1984).

According to several authorities in the management of low back pain, Waddell, (1987), it has been observed that 90% of low back injuries recover within six weeks. Farfan and Gracovetsky, (1986), determined that full ligamentous injuries require six weeks to six months for recovery. Because of these findings, it is obvious that low back injuries cannot be ligamentous in nature no matter how painful. If not ligamentous, these injuries must be musculature in nature (discounting obviously radioscopically skeletal injuries). Beiring-Sorenson, (1984),

believed that the back extensor muscles are good predictors for the future recovery of low back injury.

This tends to support the opinion that a flexed posture lift, relying on the ligamentous mechanisms might be a safe way to avoid primary and recurrent low back injury. This is especially true for workers performing prolonged activities in the squat position or having to perform frequent lifting.

Also, Sullivan, (1989), stated that the possibility of lifting an object which has no handles from the floor, requires flexion in the spine. No investigation has thus far demonstrated this can be performed otherwise.

This previous study seems to reinforce the view that the lumbar spine, due to increasing age and externally imposed mechanical restrictions such as decreased dorsiflexion, would have to call upon the flexion component. The individual would then have to resort to a squat lift (with extension) so as to not jeopardize the erector spinae and multifidus by causing repetitive excessive tensile forces. In addition, a lift requiring an extension component with the load in front of the body would be detrimental to a person having sustained an intradiscal lesion - McGill, Norman, (1985) - as it might be sufficient to cause disc failure. The mechanical limits imposed by decreased ankle dorsiflexion and the natural process of decreased upper spinal flexion in the male as described by Burton and Tillotson, (1988), would cause the person to resort, most times, to a lift using only a muscular system and would, therefore, increase the risk of soft tissue injuries most often seen clinically. Dolan et al, (1994), in their study of 'passive' extensor

moments during lifting, have determined that between 16% - 31% of the peak extensor moments are unrelated to EMG activity. If not EMG related, then peak extensor moments must be ligamentous / disc / fasciae related.

Gilad et al, (1989), reported that Anderson, Chaffin and Tichauer agreed that small changes in the spinal column configuration while lifting can cause major changes in spinal column forces.

Electromyography:

Though lifting has been studied for years, Anderson, Chaffin, Herrin et al, (1985), and the issue of lifting with a neutral lordotic lumbar spine or with the spine that is flexed, is still controversial (McGill, 1990). It is well documented and proven with electromyographic (EMG) studies of the back muscles - particularly erectors spinae, multifidus and rotators - that erector spinae muscle activity is at its minimum when flexion is at its extreme and that the erector spinae are relaxed in the initial stages of lifting. The erector spinae become increasingly more active until the upright position is achieved, regardless of the type of lift. Erector spinae EMG activity becomes very marked during straining (as in a difficult lift) and they are active on the opposite side for lateral stability when slight flexion occurs (MacConaill and Basmajian, 1977). It is for these reasons that the erector spinae muscles have been selected as being the most appropriate EMG indicator for the present study.

EMG studies of the erector spinae muscles with regard to sagittal lumbar movement and lifting has been performed by several researches such as Anderson, Herberts, Ortengren, (1976), Nachemson, (1976), and Silver, (1955). These studies implied the erector spinae contract eccentrically as the lumbar spine lowers into flexion by gravitational effect on the upper trunk.

Hart et al, (1987), reported that high EMG activity was found in the erector spinae muscles in a lift done with pelvic anterior tilt (or lordosis). However, his study did not control the distance of the load to the front of the body.

Kippers and Parker, (1984), noticed that in a normal situation, when the range of the lumbar spine approaches two thirds of flexion, these muscles are silent. The same authors believed the zygapophyseal joints have mechanoreceptors that cause a reflex shutdown of these muscles when the joint is assumed to be fully loaded. Gracovetsky, (1988), also suggested that the body calls on the most energy efficient mechanism available for a specific movement. Therefore, the natural reaction of the body is to use the thoracolumbar fasciae and midline ligamentous system as flexion would increase in the scenario where unnatural restrictions are occurring.

Nachemson, (1966), showed in his EMG studies, that the psoas major muscle act as stabilizers in the lifting process. This muscle stabilizes to avoid excessive lateral flexion and rotation. Knowing that these muscles originate on the intervertebrale discs and transverse processes of the lumbar vertebrae, Panjabi and White, (1978), and supported by Bogduk and Pearcy, (1987), found that the psoas major can adapt its roll according to the instantaneous axis of rotation of the spine. This muscle could either provide an extension moment or a flexion stabilizer depending on the starting position of the spine. No research has shown what happened after the starting position becomes altered by a mechanical disturbance.

Summary:

The human body presents a lever system designed around a compromise for speed, coordination and range of motion. With this design in mind, it is not difficult to imagine that the lifting function increases the body's risk to injury, whether from compressive forces or tensile forces on soft tissue. Much of today's manual work demands fast, repetitive lifts which could, with mechanical disturbances and increased speed, cause a serious muscular disadvantage, resulting in cumulative micro-trauma and increase fatiguability, and again, facilitating the potential for injury.

Biering-Sorensen, (1984), stated that poor endurance of back extensor muscles is a good predictor of future low back pain. However, these muscles, if forced to work in a mechanically restricted fashion, will be jeopardized even more and thereby increasing the reoccurrences of injury patterns.

Perrott, (1961), in his publication about anatomical factors relating to occupational trauma, stated that there are practical ways to avoid unnecessary fatigue and trauma (today renamed 'cumulative trauma'). His first suggestion is to eliminate unnecessary movement. "Repetition of even minute movements can be damaging". He recommended to position the body so prime movers, synergists, fixators and antagonists may each be used in the proper function (and timing).

In regards to these findings, there is a potential to modify the biomechanics at this level with natural and / or external forces. In this case, the wearing of standard safety work boots could be one of these external forces.

Therefore, consideration must be given to any cause, such as equipment or apparel, that alters this natural flow of events, thereby increasing human body risk.

METHODOLOGY

SUBJECTS:

Fifteen healthy male subjects ranging in age from 18 through to 50 years of age will be selected.

These subjects will have no previous history of injuries to ankles, knees, hips or back and will have been wearing work safety boots for a minimum of two years as a matter of daily routine at the work site.

ETHICS:

The purpose and potential significance of the research, details of what is expected from the subjects as a result of their participation in the protocol, as well as any potential contraindications as a result of their participation, will be explained to the subjects prior to their reading and signing of the University of Ottawa Ethics Committee approved (FHS-HREC) Consent Form.

EQUIPMENT AND MATERIALS:

A milk bag, plastic, bulk carrying container (Figure 1) whose outside dimensions measure 33 (L) x 33 (W) x 28 (H) mm and whose weight is 1.7 kg will be used. Four handles are built into the mold for ease of handling and it is designed

for stacking.

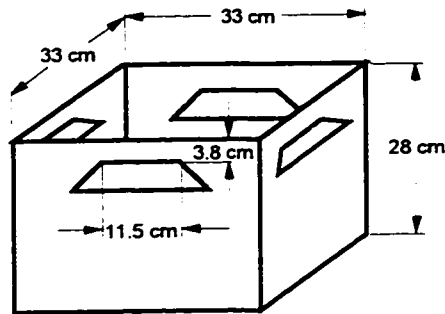


Figure 1: Milk bag carrying container

Two, serially connected, 'switch' mechanisms will be placed on the handles of the container. Their purpose is to initiate a visual and electronic time reference when the handles are grasped. A similar switch mechanism will be placed on the underside of the container to time mark the loss and re-establishment of the container's contact with the floor.

A foot placement grid (Figure 2) constructed of lines drawn 2.5 cm square will be used to have the subject's feet in approximately the same position between exercise sessions. This will be done by tracing the foot's outline with a wax pencil. The surface will be covered with a clear sheet of Plexiglas for abrasion protection and ease of cleaning.

Sufficient sand bagged ballast will be placed in the bottom of the container to bring the total weight to 11.4 kg (≈ 111 N). This is well within the 1962 ILO (International Labour Organisation) maximum limits for men (143 N) as well as being below the 'optimum condition' lift of the 1981 NIOSH 'action limit' equation (392 N) (Chaffin et al, 1984).

A Sony Video8 CCD-F501 camera on a tripod will be used to record the exercise sessions (see Figure 2 for setup) and a Nikon F-601, 35 mm camera with F: 3.5, 28-200 mm zoom lens will be used to record discrete events as appropriate.

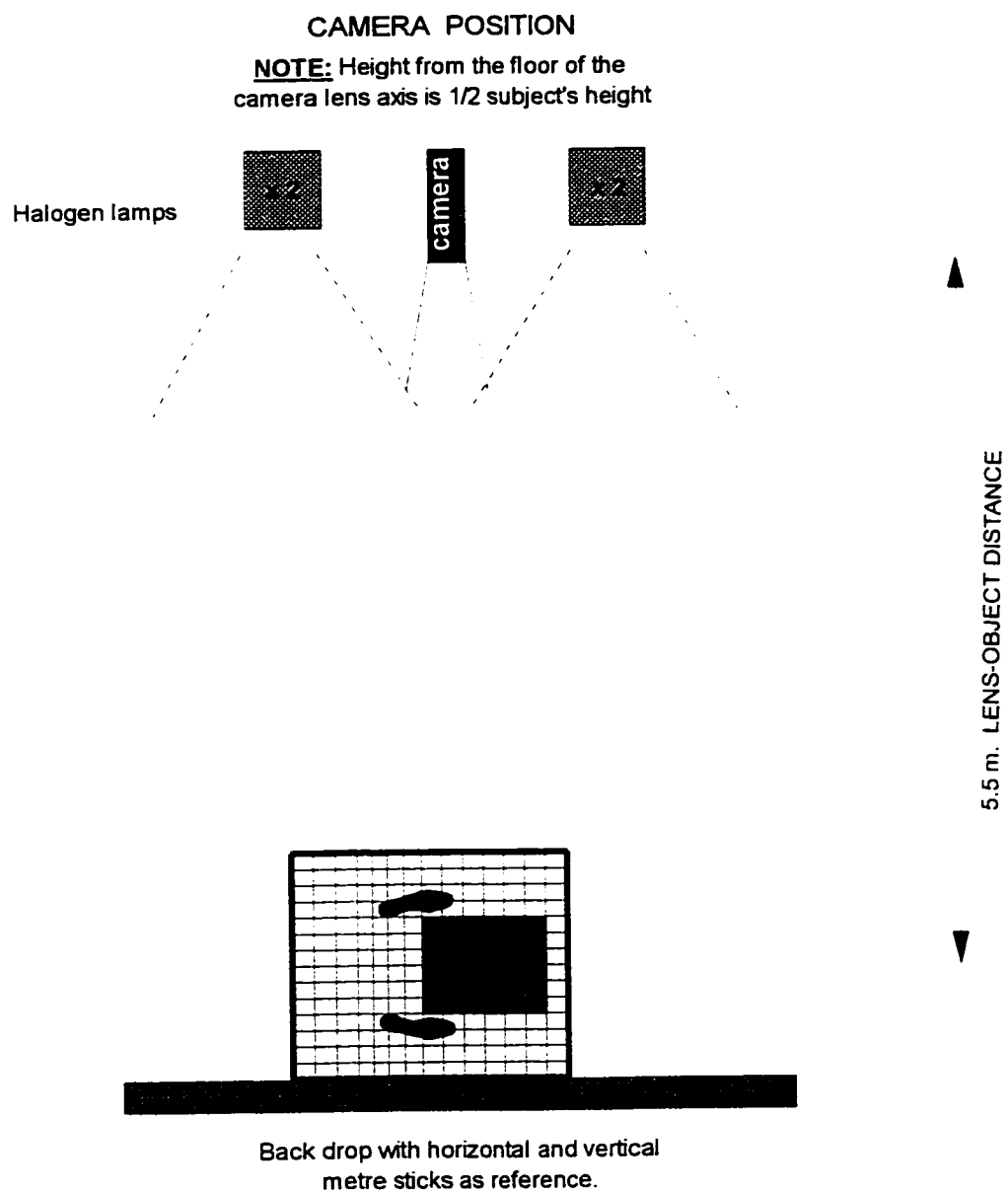


Figure 2: Elevated perspective of the basic hardware positioning for the filming sessions.

A Compaq 386 computer with VISION 3000™ software, which uses the University of Michigan's 2-D Static Model, with a 'PRESENTER®' frame-grabber board installed, will be used to examine the video tape. This computer is also equipped with the University of Ottawa's BIOAD software and a Techmar PGL40 analogue-digital signal processing board for the capturing of the EMG signals.

A 486DX2®, IBM® clone computer will be used to process the EMG signals using the University of Ottawa's BIOPROC software.

Two metre sticks mounted - one horizontally, one vertically - will provide reference points for size and position for each video frame.

Body markers of 3M® reflective tape will be used to mark the toe, metatarsal / phalangeal joint, heel, malleolus and the joint centres of the knee, hip, shoulder, elbow and wrist as well as the anterior and posterior superior iliac spine (as done by Vander Linden, 1991 and Burgess-Limerick, 1993). 'WELSH' electrodes, with the bulb painted reflective white (Figure 3) are to be used to mark the spinal reference points of S3, L3 and T10 for determining spinal flexion. AquaSonic Gel

was used to help maintain the seal for these electrodes.

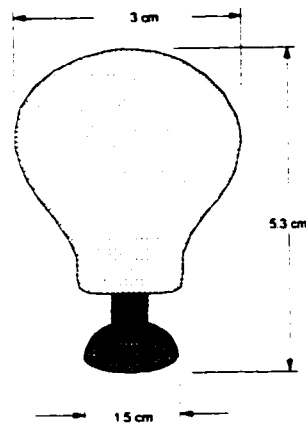


Figure 3: Quick attach (suction) 'Welsh' electrodes normally used for ECG

EMG surface electrodes (Ag - AgCl, Blue Sensor) and bioamplifiers manufactured by MEGA 3000 will be used for the 7 channels of EMG. Each pair of electrodes is to be positioned 2.5 cm apart over the specified muscles motor points.

Anthropometric measures will be taken using a leveled platform scale, a moveable anthropometre, a sliding caliper anthropometre, a spreading caliper anthropometre and an anthropometric tape measure.

A Cybex EDI 320 (Electronic Digital Inclinometre) will be used for the determination of gross Range of Motion (ROM) measurements.

PROTOCOL:

It will be explained and demonstrated to each subject what is expected of them. Consent forms will be duly signed and witnessed in compliance to the University of Ottawa Ethics Committee mandate.

The subjects will be measured according to the techniques described in the Anthropometric Standardization Reference Manual, (1988) for the following anthropometric measures : age, height, weight, and somatotype; circumference at the umbilicus, of the chest, and at the symphysis pubis; pelvis length (symphysis pubis to S1/L5); trunk length and thickness of the abdomen; lengths of shin, thigh, foot, hand, forearm and upper arm; gross ROM of ankles, hips, knees and lumbar spine. These measurements will be done on the dominant side of the body.

The skin sites for the EMG electrodes will be prepared, electrodes unilaterally attached to the longissimus thoracis, multifidus, vastus lateralis, vastus medialis, biceps femoris, tibialis anterior and gastrocnemius muscles according to the works of Delagi et al, (1975) and Warfel, (1985), and tested for connectivity to the Compaq 386. See Figures 4 and 5 for approximate electrode placements.

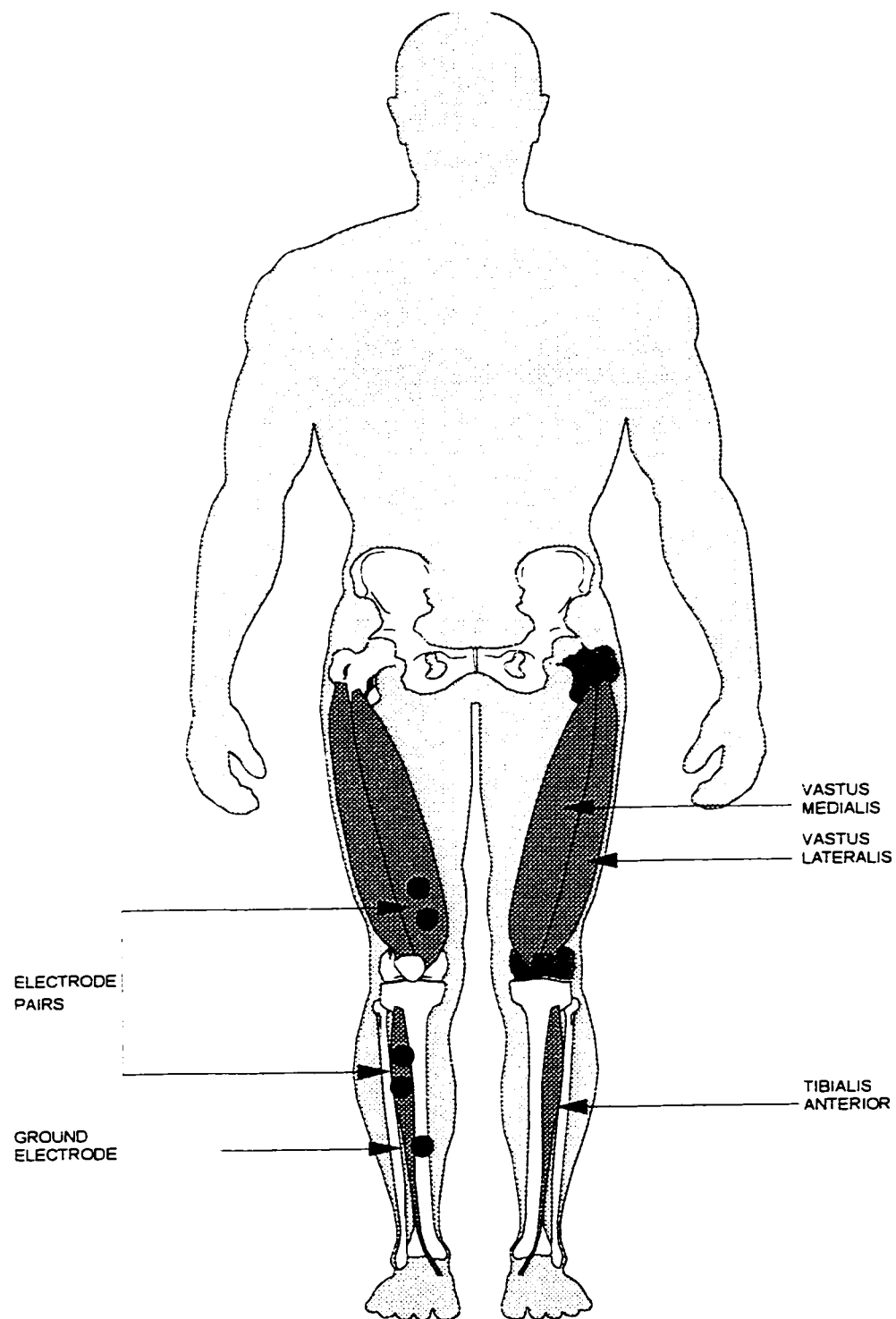


Figure 4: Approximate EMG electrode placement, anterior view.

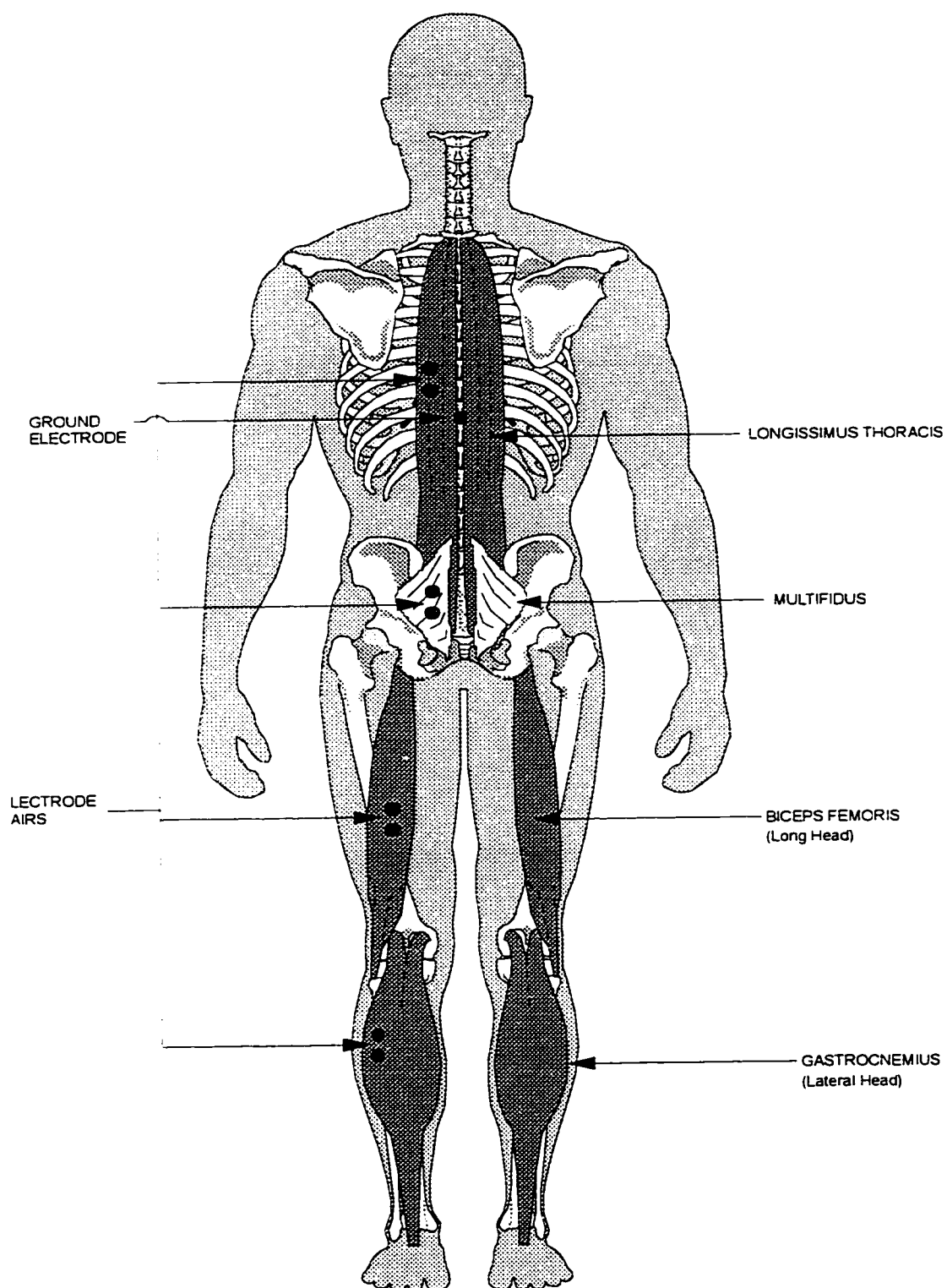


Figure 5: Approximate EMG electrode placement, posterior view.

The subjects will then be asked to approach the container and repetitively grasp, raise, lower and release it ten times, each, during two separate sessions. Ten minute rest periods will separated each session. 'Freestyle', but symmetric, squat lifts (i.e. no postural or mechanical tutoring will be given prior nor during the lifts) are to be used during each session. Each lift / lower cycle will comprised four phases: (1) the time interval between the grasping of the container's handles and the start of the container's departure from the floor; (2) the time interval between the start of the raise and the moment the subject was erect (upward movement of the container ceased), (3) time between the start of the lowering to the time the container touched the floor, (4) and, the time interval from container touching the floor to the release of the containers handles.

One session will be done barefooted (baseline for comparisons) with a heel wedge and sole identical to those of the participants workboots; the second will be done with the boots laced as the manufacturer intended them to be laced. These are to be recorded by still photography, also. The foot position on the grid of the initial session is to be marked using a wax pencil so as to maintain approximate alignment intersession. The subjects will be requested to perform the sessions in random order so as to eliminate or minimize systematic errors.

Video taping will be done at 30 frames per second, shutter speed of 1/1000th second.

EMG sampling rate will be at 500 Hz per channel based on the findings of Lamontagne and Coulombe, (1992).

DATA REDUCTION:

Electromyography

The raw EMG signals is to be recorded on the hard disc drive of the computer and later processed by 'BIOPROC' software (Lamontagne et al, 1989). The raw EMG will be AC high-pass filtered (5 Hz), bias removed using a pure moving average, full wave rectified, and low-pass filtered (5 Hz, zero-phase lag, critically dampened filter, dual pass, 4th order) to produce a linear envelope (LE). The integration of the LE-EMG will also be computed to compare areas under the curve.

The time periods of the phase 1 and 4 curves will be normalized to the time of the longest individual like phase (both intra and intersession). The time periods of phases 2 and 3 will be normalized to the longest individual phase of 2 or 3, both intra and inter session. This is done to ease comparisons of the 'lift' EMG and the 'time-reversed', 'lower' EMG. The 'time-reversed' technique is to be patterned from De Looze (1993). The global means (and other descriptive statistics) of the respective phase curves (intersession) will then be established.

Video and Anthropometry

The VISION 3000™ system provides a means of computing distances and angles directly from the video images. These features will be used to capture inter-body segment angles and distances at various stages of the lift / lower cycles; specifically the 'grasping' of the container and its 'release' after the lowering.

The 'Trunk Absolute Angle', (frontal plane, S3/T10 segment) will be computed using the technique of Chaffin et al, (1984), and Gilad, (1989), which bases all angles to the horizontal plane.

Standard descriptive statistics will be obtained from the anthropometric measures performed as part of the protocol.

STATISTICAL METHODOLOGY:

Using the GB-STAT™ statistical software from Dynamic Microsystems, Maryland, USA, repeated measures ANOVA with one factor ('Wedge' versus 'Boot') was performed on the following:

- ◆ Lumbar angle (S3-L3-T10).
- ◆ Trunk absolute angle (frontal plane, S3-T10 segment in relation to Earth's horizontal).
- ◆ Centre of mass of the trunk (T10): X-value, Y-value and average displacements.
- ◆ Ankle, knee and hip angles.
- ◆ EMG: 100% time normalized on all muscles monitored.
- ◆ EMG: peak amplitude normalized on all muscles monitored.

All of the above ANOVA's were performed on the 'grasp' moment as well as the 'release' moment.

Angles were analyzed with the ANOVA's in both degrees of arc as well as after a arcsine transformation for concurrence of significance..

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Appendix B



UNIVERSITÉ D'OTTAWA
UNIVERSITY OF OTTAWA

FACULTÉ DES SCIENCES DE LA SANTÉ
FACULTY OF HEALTH SCIENCES

and the application of a conductive gel) prior to the placing of the EMG electrodes may involve a momentary burning sensation at each of the 15 sites. Reflective body markers for the video recording will also be placed at strategic points of your foot, leg, pelvis, back, shoulder and arm. For the placement of the EMG electrodes and the body markers, a certain amount of palpation of body landmarks is required and some points may be traced directly on the skin with a pen.

The test itself will begin with a brief warm-up comprised of simple stretches. You will be asked to squat, grasp the handles of a milk crate weighing approximately 12 kg (26 lb.), raise the crate to waist height, lower it to the floor, release the handles and stand. You will be expected to do this five times in close succession. The EMG data will be recorded and you will be videoed. You will be required to repeat this sequence twice - once wearing unlaced work boots and once wearing the work boots fully laced.

Discomforts or inconveniences you may expected include: the application of the surface electrodes (EMG) and body markers for the videoing; the palpation for body landmarks; body positioning for the determination of the ranges of motion for your joints. Physical risks are minimal from this procedure.

Your privacy and anonymity will be protected in the following manner: all research data obtained about you during the course of this study will be kept confidential and accessible only to the principal investigators. Should the study be published, your identity will not be released.

This study has received approval of the Human Research Ethics Committee of the Faculty of Health Sciences of the University of Ottawa. For more information, you can contact the investigator, the advisor and the Chair of the ethics committee.

In signing this consent form, you acknowledge that you have read and understood the above statements. You enter this biomechanical investigation willingly and you may withdraw and / or discontinue your participation at any time without discrimination or penalty.

Name of Volunteer (please print):

Signature of Volunteer:

Signature of Witness :

Date:

ÉCOLE DES SCIENCES DE L'ACTIVITÉ PHYSIQUE
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OTTAWA ON K1N 6N5 CANADA

SCHOOL OF HUMAN KINETICS
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and the application of a conductive gel) prior to the placing of the EMG electrodes may involve a momentary burning sensation at each of the 15 sites. Reflective body markers for the video recording will also be placed at strategic points of your foot, leg, pelvis, back, shoulder and arm. For the placement of the EMG electrodes and the body markers, a certain amount of palpation of body landmarks is required and some points may be traced directly on the skin with a pen.

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Name of Volunteer (please print):

Signature of Volunteer:

Signature of Witness :

Date:

UNIVERSITY HUMAN RESEARCH ETHICS COMMITTEE

QUESTIONNAIRE ON RESEARCH PROCEDURES CONCERNING RESEARCH

CONDUCTED USING HUMAN SUBJECTS FOR THE HUMAN RESEARCH ETHICS COMMITTEE

FACULTY OF HEALTH SCIENCES

A typewritten response would be appreciated. Please answer all the questions.
Use additional sheets where insufficient space is provided on the questionnaire.

Name(s) of Principal Investigator(s) Michael Blench, B.A., R.M.C., P.Erg Mario Lamontagne, Ph.D., P.Erg	Department School of Human Kinetics, University of Ottawa	Office Address 125 Université Private, Ottawa, ON K1N 6N5	Telephone 564-9132; FAX: 564-7689
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1) **Title of Research Project:**
The Effect of Wearing Work Boots on Lumbar Spine Flexion
To what source is the application being submitted, if any? _____

2) **Who are the subjects (please be as specific as possible):** Healthy, male subjects 18 - 30 years of age. The subjects must have no previous history of injury to the ankles, knees, hips or back.

3) **Number of subjects involved in the study.** 15

4) **How will the subjects be recruited for this study?** The subjects will be purposefully sampled from the University of Ottawa's student body.

5) **How will you obtain the informed consent of the subject(s) (and where applicable, of parents / guardian)?** N.B. append a copy of the Informed Consent Form or Information Sheet to be used in this research. The purpose, potential significance of the research, details of what is expected from their participation, and potential contraindications will be explained prior to their reading and signing the University of Ottawa Ethics Committee approved Consent Form. See attachments.

6) **Specify the level (none, low, moderate or high) and describe the nature of the risk (legal, physical, psychological or social) associated with each major procedures with human subjects in this research. Justify the choice of this procedure(s) and state how you propose to minimize the risk.** None; lifts in the protocol are well within the NIOSH 'optimal' limit. Protocol was selected to minimize risk and fatigue.

7) **Specify the level (none, low, moderate, high) and describe the nature of the discomfort (legal, physical, psychological or social) associated with each major procedures with subjects in this research. Justify the choice of this procedure(s) and state how you propose to minimize this discomfort.** There will be no anticipated discomfort. As a result of the skin preparation for the EMG electrodes, there could be a slight, but temporary, skin irritation that could result in a slight itch.

8) **Specify the method(s) by which you plan to ensure the anonymity of the subjects and the confidentiality of the data. If you are using period data, indicate clearly how anonymity of the subjects will be protected. Where subjects are interviewed, state whether the interviewee(s) will be quoted and if so, how anonymity will be ensured. If interviewee(s) are not to remain anonymous, how will permission to quote be obtained?** Only the principal investigators will know the identity of the subjects. The research will be conducted behind closed doors and the data captured will be encrypted and in the sole possession of the investigators. 'Period data', interviews and subjects' quotations are N/A.

9) **Briefly outline what the subjects will be required to do. Indicate the number of sessions required per subject and the length of each session. Submit a copy of protocols, questionnaires or other relevant materials to be administered to subjects. Do not submit mechanical apparatus. - where scientific instruments are to be used which involve covert or overt physical contact (e.g. electrodes, sensory devices), provide a clear description of the apparatus and its function.** The subjects will be measured for basic anthropometry (15 minutes). Surface EMG electrode skin preparation will be done and electrodes placed and wired (10 minutes). While being videoed, the subjects will perform two sessions (10 seconds @) of grasping, raising to waist height (freestyle, squat lift), lowering and releasing a standard MMH container ballasted to ~ 11 kg. Five repetitions in each session with a 10 minute break between. Uni-lateral muscles for EMG: longissimus thoracic, multifidus, vastus medialis and lateralis, biceps femoris (LH), gastrocnemius (LH) and tibialis anterior. Still photography will be used as appropriate. Actual, anticipated experiment time = ~20 seconds. Subjects will be required on site for anticipated ~ 2 hours (includes setup, briefing, etc.). A complete description of the protocol is attached.

I agree to abide by the guidelines, ethical principles and the code of ethics adopted by the UHREC and its subcommittees, and where applicable, by these of the granting agency to which this proposal is being submitted, and by those of my profession or discipline, as well as by those of the facility or institution in which the research is undertaken. I am aware of my personal responsibility to be familiar with the standards. I further agree to notify the UHREC and the Human Ethics Committee of the Faculty of Health Sciences of any substantive changes in the use of human subjects in this research and to comply with requests by UHREC or its subcommittee for other information / documentation during the life of this research.

Signature of Principal Investigators:

	Day _____ Month _____ Year _____
	Day _____ Month _____ Year _____

Appendix C

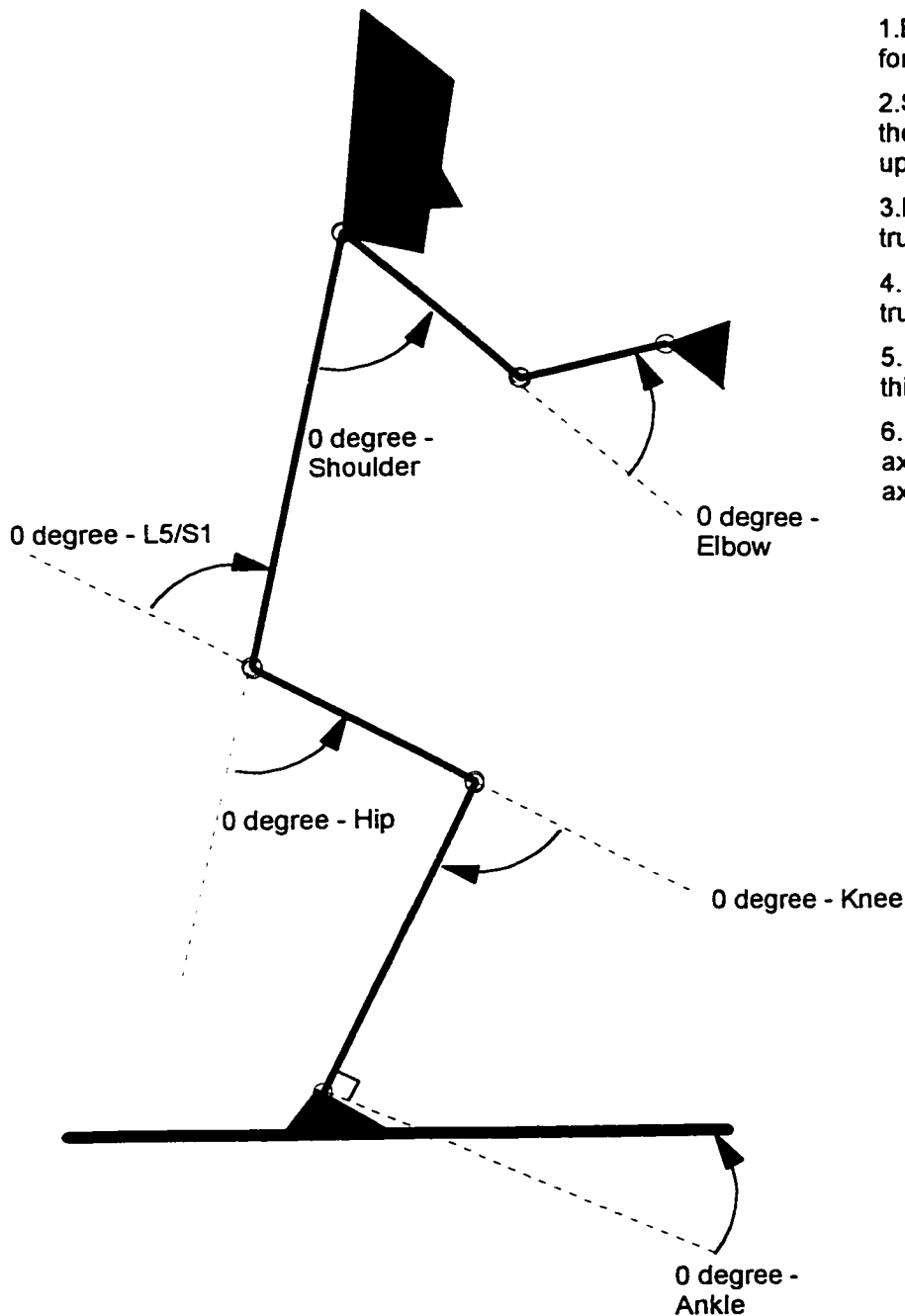
Male Anthropometrics

Subjects' Anthropometric Measures and Range of Motion																			
Subject's Number	Age	Height	Weight	Domin. Side	Somn. Type	Circ. Umbilicus	Circ. Chest	Circ. Sym.Pub. Abdomen	Thick. Trunk	Lgth. Shins	Lgth. Thighs	Lgth. Foot	Lgth. Hand	Lgth. Forearm	Lgth. Up.Arm	ROM Ankle	ROM Knees	ROM Hips	ROM L.Spine
1	39.0	173.0	78.4	1.0	3.0	92.0	98.3	93.4	22.0	41.3	46.0	26.6	18.5	27.7	38.3	77.7	124.6	106.5	47.7
2	50.0	184.6	81.4	1.0	3.0	96.0	101.5	101.0	23.0	46.5	40.0	27.6	22.0	28.5	40.8	77.7	122.3	106.7	59.0
3	28.0	176.8	68.6	2.0	2.0	85.0	94.0	98.0	17.0	42.0	42.0	27.6	22.0	28.1	37.0	61.7	119.3	84.7	61.7
4	48.0	178.6	59.3	1.0	2.0	73.5	82.5	89.0	20.5	42.0	42.0	21.6	19.0	29.2	38.7	75.4	108.3	115.5	39.0
5	33.0	182.0	100.0	1.0	1.0	101.5	110.0	108.3	27.0	42.0	44.0	27.2	22.5	26.8	37.6	68.7	114.6	106.0	62.3
6	41.0	173.4	69.0	1.0	2.0	76.5	92.0	92.2	19.0	39.5	42.7	24.4	19.2	28.9	37.3	89.0	156.0	97.0	69.0
7	27.0	176.8	73.4	2.0	3.0	83.0	96.0	98.7	20.5	43.5	44.0	26.9	21.6	28.0	37.0	83.3	118.3	128.0	69.3
8	39.0	176.7	68.2	1.0	2.0	81.6	95.0	97.0	17.5	40.4	41.0	25.6	18.6	26.0	37.2	74.3	136.7	121.3	72.0
9	54.0	194.7	89.8	1.0	3.0	90.6	107.3	101.5	22.0	43.0	48.0	28.3	21.8	29.0	40.9	70.3	120.7	122.3	60.7
10	40.0	185.5	75.7	2.0	2.0	92.8	91.5	99.0	20.5	42.0	44.5	26.2	21.0	27.5	38.4	86.3	125.3	71.0	59.3
11	24.0	188.1	74.4	1.0	3.0	83.4	89.5	92.0	20.0	41.5	47.5	27.6	21.2	28.4	38.7	75.0	108.3	110.7	61.7
12	25.0	178.1	67.7	1.0	2.0	76.0	90.0	89.5	18.7	38.2	42.0	27.2	19.6	27.3	38.2	85.0	116.7	115.7	44.7
13	19.0	171.0	77.3	1.0	3.0	83.0	94.0	103.0	19.5	38.7	44.0	26.6	18.7	27.0	35.4	86.0	125.0	111.0	79.3
14	37.0	181.6	73.6	2.0	3.0	84.8	90.5	94.0	20.6	41.5	44.0	27.5	21.6	28.0	37.4	84.3	130.7	112.3	64.7
15	34.0	188.4	107.7	1.0	1.0	100.0	112.5	113.0	24.5	43.4	45.7	28.8	22.9	28.4	38.9	77.0	110.3	113.7	63.7
16	42.0	186.6	112.5	1.0	1.0	113.8	115.0	115.2	28.5	40.0	42.0	27.3	22.6	28.4	38.9	79.3	126.7	117.0	75.7
17	27.0	183.7	78.9	1.0	3.0	82.5	98.0	100.5	20.0	39.0	44.5	27.4	20.9	28.0	38.1	79.0	129.3	114.3	62.7
Averages	35.6	181.0	79.8			88.0	97.5	99.1	21.2	41.4	43.8	26.9	20.8	28.0	39.2	78.2	123.2	108.9	61.9
						3 = eudomorphic													
						6 = ectomorphic													
						8 = mesomorphic													
Std Dev.	9.7	6.3	14.7			10.5	9.0	7.6	3.1	3.4	2.2	1.7	1.6	0.9	1.3	9.0	11.5	13.8	10.5
Var.	93.2	39.8	214.7			109.4	80.9	57.5	9.5	11.3	4.8	2.9	2.3	0.8	1.8	80.6	131.4	189.7	108.4

Appendix D

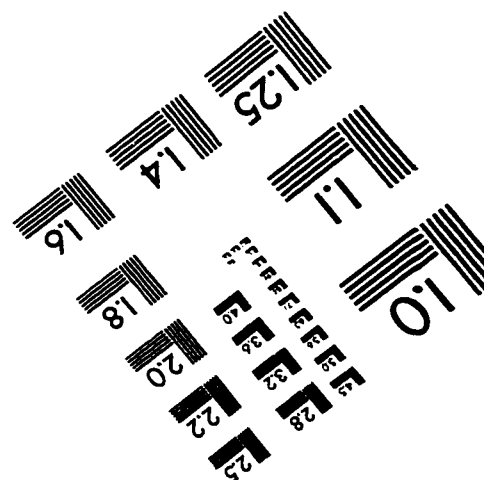
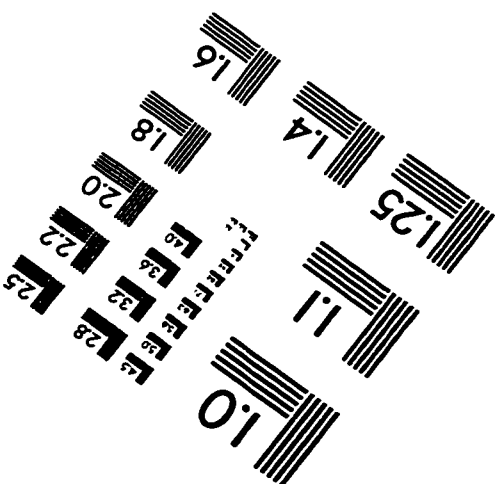
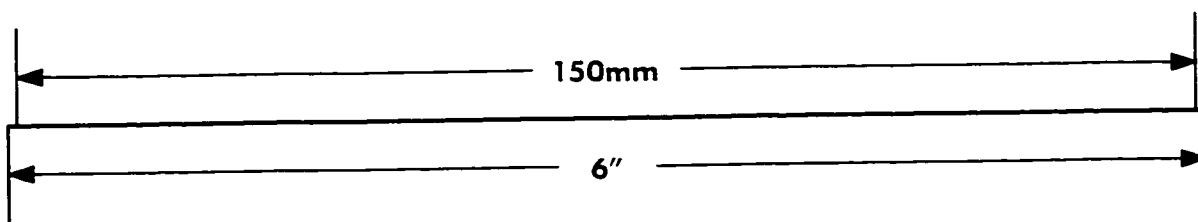
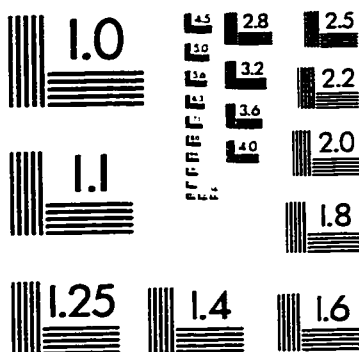
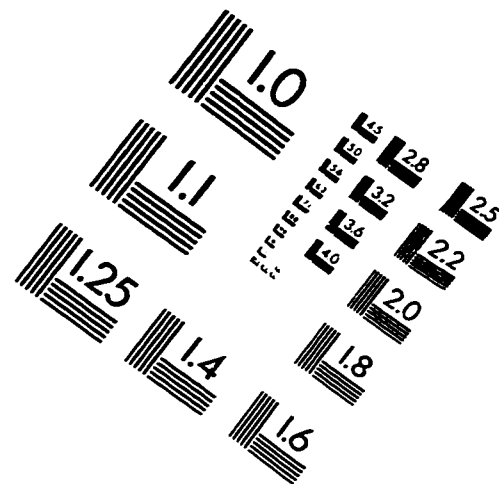
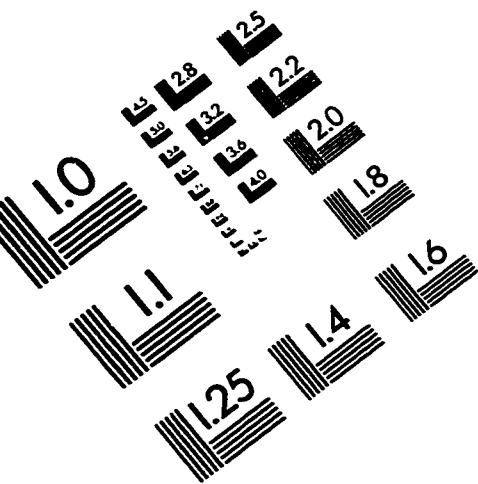
APPENDIX: D

The following diagram illustrates the goniometric reference points for the computation of relative intersegmental angles. The 'stick figure', below, has all body segments in some degree of flexion (except the foot which is in extension - dorsi-flexion) for illustrative purposes.



1. Elbow is at 0 degree flexion when the forearm is inline with the upper arm.
2. Shoulder is at 0 degrees flexion when the upper arm is coincident with the upper trunk (sagittal plane).
3. L5/S1 flexion is at 0 degrees when the trunk and thigh are inline (sagittal plane).
4. Hip has 0 degree flexion when the trunk and thigh are inline (sagittal plane).
5. Knee is at 0 degree flexion when the thigh and lower leg are inline.
6. Ankle is at 0 degree flexion when the axis of the foot is at 90 degrees to the axis of the lower leg.

IMAGE EVALUATION TEST TARGET (QA-3)



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