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Spine Loading During Whole Body Free Dynamic Lifting

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AAE	Average Absolute Error
AL	Acton Limit
DHHS	Department of Health and Human Services
EMG	Electromyography
LBD	Low Back Disorder
LMM	Lumbar Motion Monitor
LSPM	L5/S1 Position Monitor
MMH	Manual Materials Handling
MPL	Maximum Permissible Limit
MRI	Magnetic Resonance Imaging
MVCs	Maximum Voluntary Contractions
NIOSH	National Institute for Occupational Safety and Health
PAM	Pelvic Angle Monitor
PSS	Pelvic Support Structure

INTRODUCTION

Magnitude of LBD

Low back disorders (LBDs) have been described as the most common and significant musculoskeletal problem in America (Hollbrook et al., 1996, Praemer et al., 1992). Evidence of this can be seen by the fact that up to 56% of workers are seriously affected over the course of their careers (Rowe, 1981) and the estimated 70 million people in the United States who have already experienced a LBD (Cailliet, 1995). One 1992 estimate predicted an additional 7 million people will develop a LBD annually (Ayoub, 1992). In 1998, 22 million LBD cases were reported (Guo, 1993). Estimates of occupational LBD prevalence vary from 1 to 15% annually (Kelsey and White, III, 1980). Thus, LBDs affect a substantial number of individuals during their working life.

Furthermore, there are enormous costs associated with LBDs. LBDs represent the leading cause of activity limitation in people under 45 (Andersson, 1991). Half a billion lost workdays in 1988 were attributed to LBDs (Guo, 1993) resulting in $4 billion in lost wages (Frymoyer et al., 1983). Estimates of these indirect costs exceed $100 billion annually (National Institute for Occupational Safety and Health (NIOSH), 1997). LBDs account for a disproportionate portion of workers' compensation costs as well. LBDs account for less then one fifth of all worker compensation claims yet are accountable for 33-41% of costs (Hatze, 1977, Spengler et al., 1986, Webster and Snook, 1994). Additionally, indirect costs due to LBD result from worker replacement and retraining costs, loss of productivity, decreased quality and inefficiency costs. Hence, it is apparent that LBD is a tremendous burden on the workforce, both economically and in pain and suffering.Tichauer, 1976

Risks of LBD are associated with industrial work

The epidemiologic literature has noted that the type of work involved in an occupation is closely associated with the risk of suffering a LBD (Andersson, 1981, Pope, 1993). In particular, manual materials handling (MMH) activities dominate occupationally-related LBD risk (Snook et al., 1978, Bigos et al., 1986). Despite the risks involved with MMH, industrial workplaces continue to use manual labor because it remains the most cost effective method of material transfer (Drury et al., 1989). With the current growth of online services, MMH will continue to increase, particularly in warehouse environments.

Some researchers identify MMH itself as a risk factor for LBD (Adams and Hutton, 1980, Andersson, 1981, Granata and Marras, 1999, Kelsey et al., 1984a, Marras et al., 1995a, Marras et al., 1993, Punnett et al., 1991b, Tichauer, 1976) while others have identified specific factors within MMH as LBD risk factors. Epidemiological literature indicates that MMH tasks often require the worker to lift, carry heavy loads, bend forward and laterally, twist, maintain static postures, kneel during lifting, and repetitively perform combinations of these activities (Bigos et al., 1986, Gallagher et al., 1988, Kelsey et al., 1984b, Marras et al., 1993, Punnett et al., 1991a, Snook et al., 1978). In addition, high spinal loads have been associated with increased risk of LBD risk (Marras et al., 1993, Marras et al., 1995a, Norman et al., 1998, Granata and Marras, 2000). Thus, lifting and bending activities during MMH have been associated with LBDs.

Considerable evidence exists to suggest that lifting mechanically damages the spine. It has been assumed that back pain is discogenic and has a mechanical origin (Nachemson, 1975). Videman et al (Videman et al., 1990) confirmed the notion that LBD risk is associated with physically heavy work, such as MMH, by examining the functional spinal units of 86 cadavers whose work and LBD history were known. They found increased degeneration in the spines of those specimens who had performed physically heavy work. MRI studies have also been carried out on twins, showing higher rates of lumbar disc bulges in the twin who frequently experienced heavy lifting and bending (Battie et al., 1995).

Mechanical damage to the spine has been investigated in laboratory settings as well. Vertebrae-disk-vertebrae functional spinal units have been loaded to failure *in vitro* (Adams and Dolan, 1991, Adams et al., 1988, Adams et al., 1994a, Adams and Hutton, 1982, Adams and Hutton, 1986, Biggemann et al., 1988, Hutton and Adams, 1982, Lin et al., 1978, Liu et al., 1983, Liu et al., 1983, McNally et al., 1993, White, III and Panjabi, 1990). The results show the spine to be weakest when subjected to combinations of compression, shear and torsion – particularly when combined with bending (Shirazi-Adl and Drouin, 1988, Shirazi-Adl et al., 1986a). The contribution of bending to weakness of the spine may be explained by Adams and Dolan (Adams and Dolan, 1991), who found significant bending moments in the intervertebral disks resulting from flexion of lumbar vertebrae alone. Further support has been supplied by Adams et al. (Adams et al., 1994b), who used force transducers inserted into cadaveric intervertebral disks during bending to show increased pressure in the intervertebral disk as angle of flexion increased. Other *in vitro* studies have shown that shear forces, such as those experienced during extreme flexion, make the spinal motion segment far more vulnerable to injury than compressive loading alone (Broberg, 1983, Shirazi-Adl, 1989, Shirazi-Adl et al., 1986b). Additionally, Brinckmann and associates (Brinckmann et al., 1988) found that cyclic loading of the spine further reduces spinal tolerances. Hence, occupationally-related LBDs are often associated with spine loading resulting from lifting, especially in extremely flexed postures.

Therefore, it is very important for risk assessment tools to be able to estimate spinal loads as accurately as possible.

Review and analysis of existing methods for assessing lifting in industry

The high cost and prevalence of LBDs have lead to the development of techniques for lifting assessment. To this end, several control measures are widely used to evaluate the risk of occupationally-related LBD. The National Institute for Occupational Safety and Health (NIOSH) has developed two tools intended to provide a simply way to assess the hazards associated with MMH in industrial situations. The 1981 *Work Practices Guide for Manual Lifting* (Department of Health and Human Services (DHHS) and National Institute for Occupational Safety and Health (NIOSH), 1981), identifies those static lifting conditions that might increase spine compression and the risk of LBD. Jobs that impose less than 3400 N of compression on the disc, i.e. the action limit (AL), are considered safe. Jobs that impose over 6400 N of compression on the disc, i.e. the maximum permissible limit (MPL), are considered hazardous.

However, Punnett and associates (Punnett et al., 1991a) found that less than three percent of the jobs sampled in an automobile industry imposed compressive forces greater than the NIOSH AL. LBD incidence rates in this industry indicate much greater problems than the 3% of jobs that exceed the AL. NIOSH has published a revision of the 1981 lifting guide (Waters et al., 1993). The 1991 guide includes asymmetric lifting situations and considers the effects of handles on the object lifted. This guide also considers limits other than just compression on the spine.

To test the effectiveness of these guides in assessing LBD risk, Marras and colleagues (Marras et al., 1999a) recently compared predictions of the 1981 and 1991 NIOSH Guides to injury rates associated with 353 low-, medium-, and high-risk industrial MMH jobs. Both NIOSH guides were shown to be predictive, resulting in odds ratios between 3.1 and 4.6. However, each method was also shown to have definite strengths and weaknesses in regards to identifying low-, medium- and high-risk jobs. The 1981 Guide correctly identified 91% of the low-risk jobs but only correctly identified 10% of the high-risk jobs. The 1993 NIOSH revised lifting equation correctly identified 73% of the high-risk jobs and 55% of low-risk jobs. The 1981 Guide errors on the side of calling most jobs safe regardless of risk, whereas, the 1991 Guide errors on the side of calling most jobs dangerous regardless of risk. Thus, the current easily applicable means of controlling LBD risk that are based primarily on spine compression limits are extremely limited in their ability to assess risk. Furthermore, current ergonomic control measures unrealistically assume every individual and every lift generate the same spinal loads. Not every worker is the same, nor is every repeated exertion performed in the same manner.

Other risk assessment tools exist for evaluating industrial job risk, but they require specialized instrumentation. Marras et al (Marras et al., 1993, Marras et al., 1995a, Marras et al., 2001) have developed one such tool. Using workplace and trunk motion factors quantified during industrial jobs, they developed a multiple logistic regression model capable of differentiating between high-, medium- and low-risk (for LBD) jobs with odds rations of up to

10.7 (low- vs. high-risk jobs). The model used several used motions from several representative trials and then assessed risk based on 5 factors: 1) lifting frequency, 2) load moment, 3) trunk lateral velocity, 4) trunk twisting velocity, and 5) trunk sagittal angle. Of these factors, maximum load moment was the single most powerful predictor, with an odds ratio of 5.17. Thus, this model underscores the need for considering trunk dynamics as well as moment imposed on the spine in workplace assessments.

Biomechanical models

While trunk dynamics may be measured *in vivo* through the use of specialized goniometers, direct measurement of spinal loading is more difficult. Spinal pressures during lifting have been directly measured *in vivo* (Andersson et al., 1976, Andersson et al., 1977b, Nachemson and Elfstrom, 1970). However, because direct measurements of spinal loads are difficult to carry out, various methods have been created to quantitatively estimate spinal loading. Unlike the NIOSH guides, which are designed to identify jobs with high risks for LBD, biomechanical models are intended to estimate spinal loads. These estimates may later be used for job assessment by comparing loadings associated with different job tasks directly or by comparing estimated loads to empirically determined spinal tolerances.

Techniques for estimating spinal loading vary in complexity from simple measurements of external moment calculated from the weight lifted and the moment arm between the load and the worker (Chaffin and Baker, 1970, Marras et al., 1995a, Andersson et al., 1976) to complex models of internal spinal loading using mathematical models of spine geometry, electromyography (EMG) of the trunk musculature during lifting, trunk kinematics and estimates of the contribution of passive tissues (Marras and Granata, 1997b, McGill and

Norman, 1986, Schultz and Andersson, 1981). Since the low back is often associated with work-related LBP, these models typically focus on estimating forces on the intervertebral disc in the L4/L5 or L5/S1 joints (Leamon, 1994).

Traditionally, most biomechanical assessments of lifting situations have been limited to static evaluations of the trunk (Andersson and Ortengren, 1974, Andersson et al., 1976, Andersson et al., 1977a, Chaffin, 1969, Chaffin and Muzaffer, 1991, Chaffin and Baker, 1970, Schultz et al., 1987, Schultz and Andersson, 1981, Schultz et al., 1982). Many of these models assume that the lifted load is counterbalanced with one equivalent back muscle, and have focused primarily on spinal compression associated with lifting under sagittally symmetric conditions.

As previously noted, epidemiologic studies indicate that repetitive twisting, extreme flexion or lateral bending and lifting, even for relatively light loads, are significant risk factors for LBD indicating the importance of estimating three dimensional spine loads. These lifting postures are expected to increase shear and torsional loads on the spine. The influence of shear and torsional load in combination with compressive loads in the spine has been under-appreciated because few static biomechanical models are able to assess multi-dimensional loads.

Static biomechanical models also neglect the effects of motion of the external loads. It is known that dynamic lifting tasks in increase the magnitude and variability of spinal loads (Freivalds et al., 1984, Han et al., 1995, Lindbeck and Arborelius, 1991,Marras and Sommerich, 1991b, McGill and Norman, 1986). There is also in-vitro evidence that the viscoelastic properties of the ligamentous spine may act to increase spinal stress with increased speed of spine motion (Adams and Hutton, 1985, Andersson et al., 1974, Brinckmann et al., 1988,Broberg, 1983). Additionally, lifting velocity (dynamic lifting) has been associated with increased trunk muscle coactivity, coactive variability, and intra-abdominal pressure (Marras and Kim, 1993, Marras et al., 1986, Marras and Granata, 1995, Marras et al., 1984, Marras and Mirka, 1990). Muscle coactivity has also been shown to significantly increase spinal loads during dynamic lifting (Granata and Marras, 1995b). Therefore, spine compression during dynamic lifting will be greater than that predicted by static models due to the effects of lifting velocity on muscle coactivity (Davis and Marras, 2000b, Marras and Reilly, 1988, Marras and Sommerich, 1991a, Marras and Sommerich, 1991b, McGill and Norman, 1985). In fact, neglecting lift dynamics has been shown to under predict spinal loads during dynamic exertions by 22.5% to 60% (Marras and Sommerich, 1991a) and by as much as 90 to 250% when both trunk dynamics and muscle coactivity are neglected (Davis and Marras, 2000b).

Similar logic may be applied to shears imposed on the spine during dynamic lifting. As previously noted, shear forces make the spinal motion segment far more vulnerable to injury than compressive loading. Since shear and torsional loads become more prevalent when the speed of trunk motion increases, the effects of dynamics become even more important (Marras and Sommerich, 1991a).

Similarly, the variability of spinals load increase with lift velocity (Mirka, 1991). This indicates that a task may be associated with average spinal loads below tolerance limits for the spine, but repeated performance will generate a significant number of exertions with spinal loads greater than acceptable tolerance.

Thus, we expect that a more detailed evaluation of in-vivo occupational lifting conditions, one that includes documentation of the trunk motion characteristics in industry, may better explain the occupational source of LBD. To accomplish such an evaluation, an accurate, dynamic three-dimensional, biomechanical model will be needed.

In conclusion, it is extremely important that the biomechanical tools used to assess MMH in the workplace accurately consider the effects of free-dynamic motion during lifting. Once successfully achieved, biomechanical analyses may be performed to assess specific subject and trial variability. This will allow more accurate risk assessments and will facilitate future research into the assessment of work-shift duration, rotation, environmental variables, and psycho-social effects upon the risk of LBD.

Biomechanical model development

A biomechanical model of spine loading has been developed continuously and improved in the Biodynamics Laboratory at The Ohio State University over the past 18 years. This model has been developed to address the previously mentioned issues by taking into account trunk dynamics during lifting as well as muscle coactivity through the use of EMG monitoring of trunk musculature.

The model initially described the complex relationship between the activity patterns of the antagonistic and agonistic muscles during dynamic lifting (Marras and Reilly, 1988). Based on the temporal relationships between various trunk muscles during dynamic activity, Reilly and Marras (Reilly and Marras, 1989) developed a spinal loading model. This early model used a simple relationship for the muscle activities – estimating the EMG profile using two straight lines (triangular shape). These models were developed from data collected from male subjects using an isokinetic dynamometer. During the experiment, the subject performed lifting exertions between 0° and 45° of flexion while standing with their legs straight and hips unrestricted, as described by Marras and associates (Marras et al., 1984).

Marras and Sommerich (Marras and Sommerich, 1991a, Marras and Sommerich, 1991b) enhanced the development of the biomechanical model by normalizing the trunk muscle activity as a function of trunk position and modulation factors for muscle length-strength and force-velocity relationships to estimate spinal loading. Additionally, the EMG activities were estimated more accurately through the use of three linear estimations (a trapezoid). During the experiment, male subjects performed simulated sagittally symmetric and asymmetric lifting exertions between 0° and 45° of flexion while standing with their legs straight and hips restrained. Validation of the model consisted of comparing the predicted moments to the measured moments, where R-squares were greater than 0.7 over 85% of the time.

Granata and Marras (Granata and Marras, 1993) further developed the model to incorporate instantaneous EMG instead of linear estimations based on a few EMG data points. This model also used the gain (representing the subject's muscle strength per unit area) as a subject-dependent constant. Model performance was good with 80% of the trials having R^2 values greater than 0.80, and an average gain of 42 N/cm^2.

The model has been adjusted further to incorporate the artifacts of lengthening of the trunk muscle while exerting, such as during a lowering task (Davis et al., 1998). The performance of the model for eccentric contractions during lowering tasks with the subject's hips restrained was better than for concentric lifting with most R^2 values greater than 0.95.

Recently, the model has been applied to more realistic, free-dynamic lifting situations by removing the hip restraint used in previous studies. The LSPM and PAM goniometers developed in this study were used to assess free-dynamic lifting (Granata and Marras, 1995a). Subjects stood on a force plate with the goniometers attached and their hips unrestrained. They were instructed to lift with their legs straight but their hips were allowed to rotate. The results of this study indicated that the unadjusted model was able to perform well free-dynamically with an average R^2 value of 0.82, average gain of 64.9 N/cm^2 and average AAE of less than 15% of measured sagittal moment. However, the decrease in R^2, increase in gain and increase in AAE when compared to lifts with the hips secured indicated that some modification to the model will be beneficial to adjust it for whole body free-dynamic lifting.

The model has also been adjusted for the differences between female and male muscle anatomies (Marras et al., 2001, Jorgensen et al., 2001). MRI scans of 20 female subjects were used to quantitatively describe the internal geometry of the female trunk musculoskeletal system so that the model accurately represents internal trunk mechanics. At the same time, MRI scans were taken of 10 male subjects and the internal trunk mechanics of the original, male model was updated accordingly. Using the female-specific internal geometry, female-specific length-strength and force-velocity relationships were established. These models were then applied to free dynamic lifting tasks performed by male and female subjects. As with Granata and Marras (Granata and Marras, 1995a), subjects were instructed to lift with their knees straight but their hips were not restrained. The mean R^2 values for males and females, respectively, were 0.87 and 0.86. Mean gains were 56.1 and 54.0 and average AAEs were 9% of measured moment for both genders. Thus the model can also be used to provide information about differences between genders.

This model has been progressively validated for sagittal bending (Granata and Marras, 1993, Granata and Marras, 1995a, Marras and Sommerich, 1991a, Marras and Sommerich, 1991b, Mirka and Marras, 1993), lateral bending (Marras and Granata, 1997b), and axial twisting (Marras and Granata, 1995).

The purpose of this study was to continue the model's evolutionary process by considering the effect of pelvic tilt on spine loading during realistic lifting tasks. At the time this study began, the apparatus did not exist to use the model for free-dynamic lifting. Thus, the first goal of this study, to create goniometers to allow the model to be used to assess free-dynamic lifting, has already been put into use (Marras et al., 2001, Granata and Marras, 1995a). The next goals of this study were directed towards applying the model to an increasing range of realistic lifting situations. Thus, this study focused on adjusting the model to include whole body free-dynamic lifting so that subjects will no longer be required to keep their legs straight during lifting while at the same time showing the model's validity in extreme postures and with knee bending.

METHODS

Study Approach

There were three objectives for this study. The first objective was to build a goniometer system to instantaneously monitor the location of the lumbo-sacral joint, pelvic orientation, hip motion and knee motion. A L5/S1 position locator and a pelvic angle goniometer were constructed to allow mathematical translation and orientation of forces and moments recorded via a forceplate under the subject's feet to the lumbo-sacral joint (Fathallah, 1995). In addition, goniometers for the hip and knee were constructed to determine changes in length of the trunk musculature due to lower extremity postures.

The second objective was to adjust the biomechanical model for whole body free-dynamic lifting by using these goniometers to systematically determine a correction factor for erector spinae muscle length in the biomechanical model as pelvic tilt changes. In effect, this correction factor adjusted the length-strength relationship within the model. It was expected that changes in pelvic tilt, measured through external sagittal hip angle, would affect the length of the erector spinae muscles due to their attachment points on the pelvis. Early testing highlighted difficulties directly controlling hip angle during free-dynamic lifting. Subjects found it extremely difficult to control hip angle, even with the aid of direct visual feedback presented from the hip monitor. In addition, analysis of data collected while subjects were required to directly control hip angle showed decreased sagittal range of motion, increased coactivity, and in increased subjective fatigue ratings. For these reasons, hip angle was not directly controlled during lifting. To provide some control over hip angle, subjects' knee angle was directly controlled instead. The rationale for this was that controlling knee angle indirectly controls the angle of the hips and trunk required to maintain balance. Trunk angle was also controlled during lifting so the only variable the subject could adjust to compensate for the set knee angles was the angle of the hips.

The third objective was to validate the adjusted model under whole body free-dynamic lifting conditions while controlling as little of the lift as possible. Therefore, during this part of the study, the previous controls on trunk and knee angles were relaxed and subjects were instructed to lift using either a stoop (straight-legged) or squat (bend the knees) lift, depending on the experimental condition. The goal of this phase was to verify that the adjusted model performs well under realistic lifting conditions.

Subjects

All subjects were recruited from The Ohio State University student body and the local community. Twenty male and 20 female volunteers participated in the model adjustment phase (Part Two). The model validation phase (Part Three) consisted of 18 male and 18 female volunteers. All subjects had been asymptomatic of LBD for at least one year prior to participating in the experiment. Subjects in all studies had similar anthropometric dimensions. Subject anthropometry is shown in Table 2.1.

Table 2.1: Subject Anthropometry. Standard deviations in parenthesis.

	Part II		Part III	
	Males (20)	Females (20)	Males (18)	Females (18)
Age (years)	23.1 (3.2)	23.8 (5.0)	23.3 (4.7)	22.4 (3.6)
Weight (kg)	74.0 (19.6)	61.1 (7.3)	78.1 (11.9)	61.3 (9.9)
Standing Height (cm)	179.0 (8.3)	158.8 (34.6)	178.4 (11.2)	164.8 (6.0)
Shoulder Height (cm)	149.0 (11.2)	137.5 (5.5)	148.6 (7.5)	137.2 (5.5)
Elbow Height (cm)	110.6 (6.1)	102.0 (4.4)	108.6 (11.6)	101.2 (4.1)
Upper Leg Length (cm)	48.2 (18.1)	49.4 (14.0)	48.0 (17.4)	42.5 (4.8)
Lower Leg Length (cm)	54.2 (3.4)	51.6 (3.1)	55.1 (4.3)	49.6 (3.2)
Upper Arm Length (cm)	37.2 (4.2)	34.9 (2.7)	37.5 (2.8)	35.0 (1.6)
Lower Arm Length (cm)	47.3 (3.2)	43.2 (1.9)	46.9 (6.2)	44.1 (1.7)
Spine Length (cm)	55.1 (13.8)	51.0 (4.4)	57.2 (7.5)	51.1 (3.5)
Trunk Depth (pelvis) (cm)	21.7 (3.2)	18.2 (2.1)	21.3 (2.5)	18.8 (3.1)
Trunk Breadth (pelvis) (cm)	28.0 (4.2)	27.2 (2.4)	29.4 (3.5)	27.1 (3.4)
Trunk Depth (xyphiod) (cm)	22.5 (3.0)	19.1 (1.5)	21.5 (1.9)	18.7 (1.7)
Trunk Breadth(xyphiod) (cm)	32.0 (9.7)	27.4 (2.8)	30.8 (2.4)	29.7 (12.0)
Trunk Circumference (cm)	83.3 (7.6)	76.9 (11.2)	82.5 (12.7)	74.8 (6.1)
%Right-Handed	85	95	94	94
% Caucasian	65	70	61	65
% Asian	20	10	11	12
% Hispanic	05	05	00	00
% Black	00	05	06	06
% Other	10	10	22	18

Experimental Design

Part One: Development of Instrumentation to Allow for Whole Body Free-Dynamic Lifting in the Model

Part One was a systems development and building project. One set of goniometers was developed to translate and orient forces and moments measured at the force plate to align with the subject's lumbo-sacral joint. Another set was developed to monitor the position of the subject's hip and lower extremity in order to determine possible changes in muscle length due to hip flexion during lifting.

Both sets of goniometers are shown on a subject in Figure 2.1.

Figure 2.1: Experimental setup showing goniometers used to monitor body kinematics.

Part Two: Adjustment of the Biomechanical Model for Whole Body Free-Dynamic Lifting

Independent variables consisted of: 1) four static knee angles (Figure 2.2) maintained while subjects performed trunk extensions from a 45° sagittally flexed position to an upright posture, 2) three trunk extension velocities, and 3) two load (weight) conditions.

The experimental knee positions consisted of four equally spaced knee positions as shown in Figure 2.2. Knee position was used as an indirect control of hip position so that hip positions expected throughout a full squat lift could be tested. A review of video recorded in industrial settings (Marras et al., 1993) has shown that these squat lift knee angles also include all knee angles observed during stoop lifting.

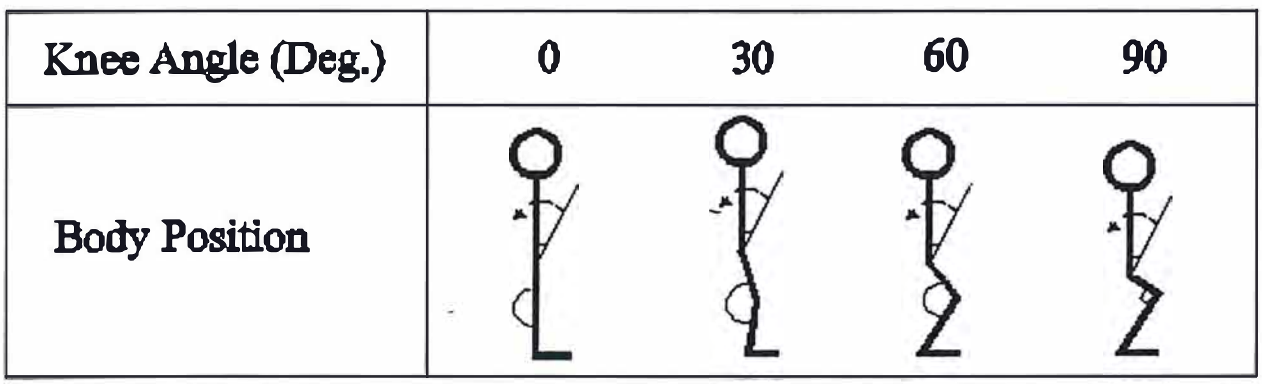

Figure 2.2: Lower extremity postures tested for derivation of the trunk muscle length correction factor.

Three experimental free-dynamic trunk extension velocities, 20, 40, and 60°/sec, were studied in this experiment. These values approximate the low, medium, and high-risk trunk velocities observed by Marras and colleagues (Marras et al., 1995a). The final independent variable, load weight handled during the lift, was set at two levels. Subjects lifted weights of 6.8 and 13.6 kg, which represent the weight range of observed low- and medium- risk (of

LBD) industrial loads (Marras et al., 1995a). Subjects were required to repeat each experimental condition two times, presented in counterbalanced order.

Part Three: Validation of the Adjusted Model Under Realistic Lifting Conditions

The dependent variables in the model validation phase consisted of model measures of performance – gain, R^2 and average absolute error (AAE). Accuracy of estimation for changes in trunk loading during the trials was measured using R^2 correlations between the measured trunk moment and predicted trunk moment over the entire trial. The accuracy of magnitude of the predicted load imposed upon the trunk (moment) was assessed by comparing the RMS error between the measured trunk moment and the predicted trunk moment during a lifting trial (AAE). Predicted muscle gains (force predicting capability per unit area) were also analyzed as a measure of model feasibility.

Independent variables consisted of: 1) two lifting styles, 2) two lift frequencies, 3) two lifting asymmetries, 4) two lift origin heights and, 5) two load magnitudes.

The lifting styles consisted of a stoop lift where subjects were instructed to perform the lift without bending their knees and a squat lift where subjects were instructed to perform the lift as they wished including, but not requiring, bending the knees. These two styles represent the extremes of lifting styles. Most lifting involves some combination of these two styles (Splittstoesser et al., 2000).

Lifting frequency is intended to solicit different lifting speeds without directly controlling lifting velocity. Lifting frequency was set at 2 lifts per minute and 8 lifts per minute. These conditions represent a relatively slow as well as a fast lifting condition. These conditions also represent the extremes of the range of lift frequencies observed in industry (Marras et al., 1995a).

Lifting asymmetry consisted of a sagittally symmetric lifting condition, and a 45^0 clockwise asymmetric lifting condition. Studies have shown that 45^0 of asymmetry is approximately the limit of twisting of the body before a worker compensates by taking a step (Marras et al., 1995a). Since subjects were required to keep their feet stationary during lifting, this asymmetry would be applicable to the more extreme task asymmetries found in industry.

Vertical lifting height conditions consisted of 1) a starting height at ankle level and 2) a starting height at knee level. These two starting heights were considered to represent an adequate range of hip and pelvic rotation that is commonly seen in industry.

Load magnitudes consisted of 6.8 and 13.6 kg which represent the weight range of observed low- and medium- risk (of LBD) industrial loads as well as being the same loads used in this experiment to adjust the biomechanical model for whole body free-dynamic lifting (the model adjustment phase) (Marras et al., 1995a).

Apparatus

This section outlines the devices used throughout all three phases of this study. The L5/S1 Position Monitor (LSPM), Pelvic Angle Monitor (PAM), Hip Monitor and Knee Monitor were all constructed during the goniometer development phase. All of the devices described in this section were subsequently used during the model adjustment and validation phases.

A Lumbar Motion Monitor (LMM), a triaxial goniometer capable of collecting trunk sagittal, lateral and twisting motions simultaneously, was used to collect trunk kinematics

(Marras et al., 1992). The LMM was placed on the subject's back and secured through orthoplasts around the waist and thorax. The LMM is shown in Figure 2.3.

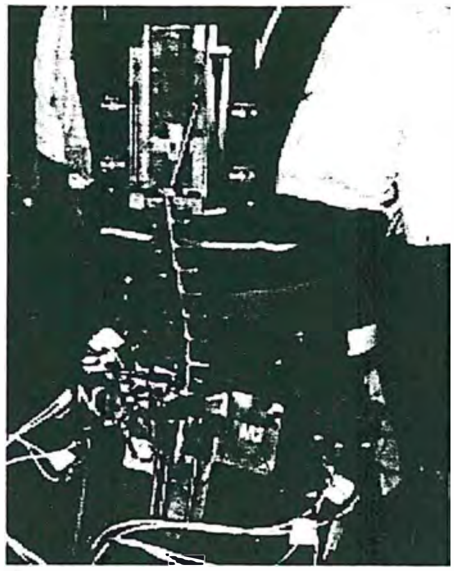

Figure 2.3: The Lumbar Motion Monitor

A three dimensional L5/S1 joint position locator was built to continuously locate the subject's L5/S1 position relative to the forceplate. The LSPM was located at a point 80 cm behind and 80 cm above the force plate and was attached, via a thin retractable lead, to the PAM affixed to the LMM over the subject's L5/S1 joint. Details of the construction, calibration and application within the model of the LSPM are given in the Results section. Fathallah and Granata and associates have also reviewed the LSPM Fathallah, 1995, Granata et al., 1996. The LSPM is shown in Figure 2.4.

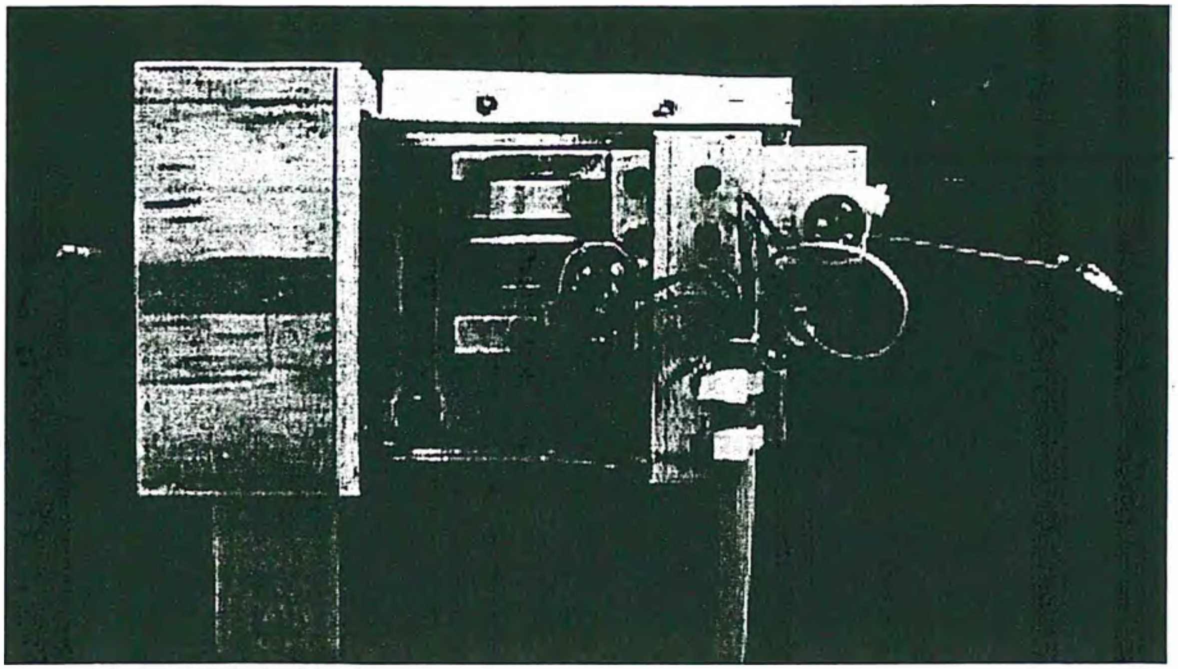

Figure 2.4: The L5/S1 Position Monitor.

The PAM was constructed to accurately monitor pelvic orientation relative to the force plate. The PAM connects to the retractable lead from the L5/S1 locater via a guide and uses a pair of potentiometers mounted to the guide to continuously monitor the horizontal and vertical angles between the LSPM's retractable lead and the subject's pelvis. The PAM is shown in Figure 2.5. Details of the construction, calibration and application within the model of the PAM are given in the Results section.

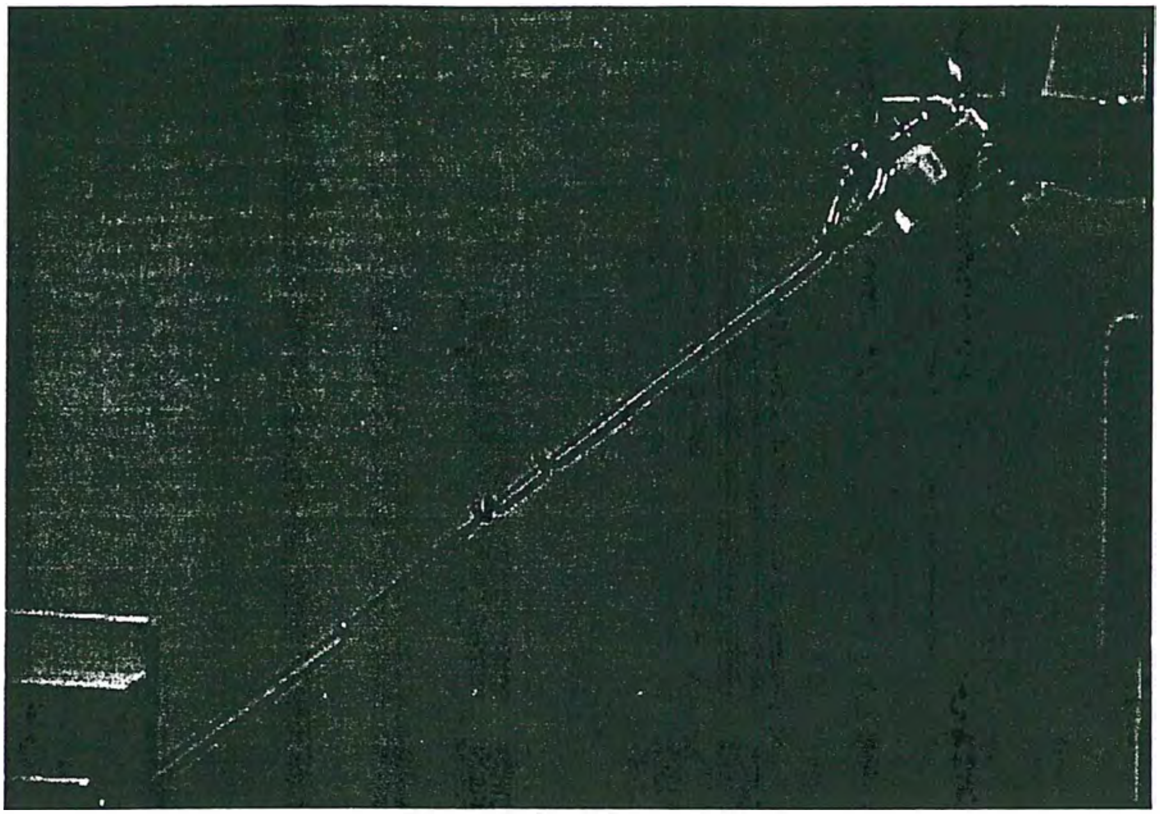

Figure 2.5: The Pelvic Angle Monitor.

The Hip Monitor continuously monitored the position of the upper leg relative to the pelvis in flexion-extension and abduction-adduction. The Hip Monitor consisted of a rod attached to the subject's thigh and pelvis via orthoplasts. At the pelvis, the rod connects to a universal joint placed over the center of rotation of the hip joint. A pair of potentiometers continuously monitor the motion of the universal joint in flexion-extension and abduction-adduction of the hip as the subject moves. The Hip Monitor is shown in Figure 2.6. Details of the construction and calibration of the Hip Monitor are given in the Results section.

Figure 2.6: The Hip Monitor on a subject.

A Knee Monitor was used to continuously monitor the subject's knee angle. The Knee Monitor consists of two thin metal strips with a joint in the middle. A potentiometer allows monitoring of the angle between the two strips. This design has been previously used to monitor wrist joint angles during industrial and keyboarding tasks (Schoenmarklin and Marras, 1993). To keep the Knee Monitor in place during lifting, subjects wore a hinged knee brace on their right leg (Donjoy Hinged Knee Waparound, Vista, CA). This knee brace was chosen because it was a sport knee brace, designed to remain in place during physical activity and also to provide lateral support without affecting sagittal motion. Thus, the knee brace provided an excellent mounting place for the Knee Monitor. The Knee Monitor was positioned at the center of rotation of the subject's knee joint and held to the knee brace using the knee brace's Velcro strapping and tape (if necessary). The Knee Monitor is shown in Figure 2.7.

Subjects performed the lifting exertions while standing on a force plate (Bertec 4060A, Worthington, OH), which measured the three dimensional ground reaction moments and forces generated during the lifting exertions.

Electromyographic (EMG) activity was collected by bipolar silver-silver chloride surface electrodes, spaced approximately 3 cm apart over ten trunk muscles (Marras and Mirka, 1993). The ten trunk muscles included the right and left pairs of the latissimus dorsi, erector spinae, rectus abdominis, external obliques, and the internal obliques. The EMG data was notch filtered at 60 Hz, high- and low-pass filtered at 30 and 1000 Hz respectively, rectified and integrated using a 20 ms sliding window hardware filter (Marras and Mirka, 1993).

Figure 2.7: The Knee Monitor on a subject.

All data signals from the above equipment were collected at 100 Hz simultaneously through customized Windows™ based software developed in-house. The data was recorded on a Pentium computer via an analog-to-digital conversion board and stored for later analysis.

In both phases, weights lifted were placed in a 28 cm by 30 cm by 22 cm wooden box with handles located at the centers of each side. The box was handed to the subject after they had achieved the start posture and before the beginning of the lift. A typical experimental setup is shown in Figure 2.8.

Figure 2.8: A typical experimental setup. Knee/Hip Monitors on far side.

Procedure

The following procedure pertains to the model adjustment and validation phases of the experiment. Portions of the experimental procedure that differ between parts are explained below. Upon arrival to the testing laboratory, the subjects read and signed a consent form. Anthropometric data and demographic information were recorded next. Surface electrodes were then applied over each of the ten trunk muscles and skin impedances were kept below 500 kΩ. Next, static maximum voluntary contractions (MVCs) for each of the trunk muscles were obtained. Subjects were placed into a rigid structure and static MVCs performed in 6 directions: trunk extension with the trunk sagittally flexed 20°, flexion, right and left lateral bending as well as clockwise and counterclockwise twisting exertions (Marras and Mirka, 1993). After each maximum exertion, two minutes of rest was given to reduce the effects of fatigue (Caldwell et al., 1974). All resulting trunk muscle EMG data obtained from the experimental trials were normalized relative to the maximum EMG activity obtained for each muscle during these six directional MVCs. Following the MVCs an LMM was placed on the subject's back, the PAM, LSPM, Hip Monitor and Knee Monitor were all attached to the subject as they stood upon the force plate. The Hip and Knee Monitors were attached to the subject's right side. After the goniometers were in place, the subject was allowed to practice the lifting motions to become proficient with the different experimental conditions.

Part Two: Adjustment of the Biomechanical Model for Whole Body Free-Dynamic Lifting

The experimental task required the subjects to simultaneously maintain set knee angles while controlling their trunk lifting velocity between tolerance limits. During the lift, subjects viewed a computer display of the instantaneous angle of sagittal flexion reported by the LMM and a numerical display of the Knee Monitor angle in real time. Subjects controlled their isokinetic extension velocity by keeping the displayed trace of the instantaneous trunk sagittal position within a tolerance region displayed on the computer. A five percent tolerance was used for velocity by displaying two lines that were 2.5% around the target velocity. Knee angles were required to stay within 5° of the set level (Figure 2.9). If the subject failed to maintain the trunk motion or knee angle, the trial was repeated. Each of the trials was presented to the subject in a randomized order.

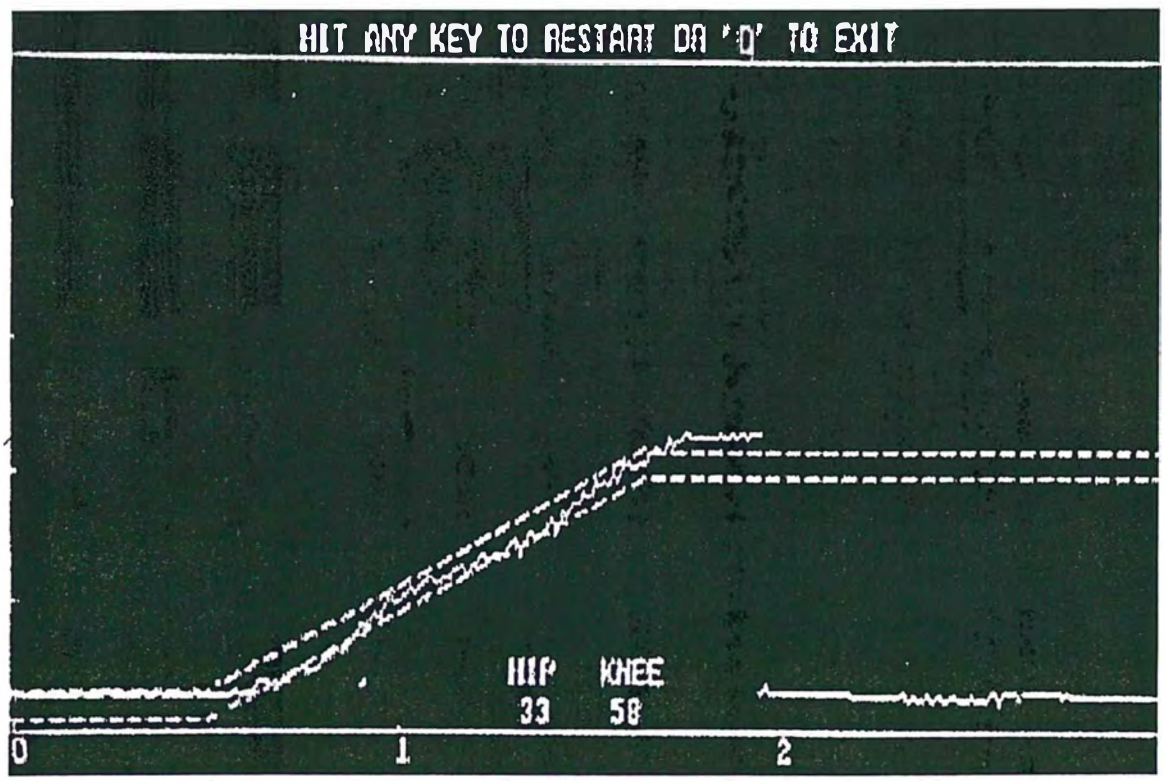

Figure 2.9: Biofeedback display used by subject to control trunk motion and knee angle.

Part Three: Validation of the Adjusted Model Under Realistic Lifting Conditions

After the goniometers were in place, the subject was allowed to practice the lifting techniques to become comfortable with stoop and squat lifting styles. The subject performed the lifts in sets of three – lifting at the sound of a tone triggered by computer timer. The first lift of each set was discarded to reduce the effects of previous lifting rates on the subject's style. Each of the experimental trials was presented to the subject in a randomized order and subjects were instructed to keep their feet stationary during each experimental condition. The lifting techniques were explained to the subject as follows: for stoop lifts, the subject was instructed to keep their legs straight throughout the lift; for squat lifts, the subject was told to bend their knees as much as they wished but no exact amount of knee flexion was specified. The subject was instructed to begin each lift at the sound of the computer-generated tone and remain stationary (standing) between the end of the lift and the following tone.

Data Analysis

Part Two: Adjustment of the Biomechanical Model for Whole Body Free-Dynamic Lifting

The model was originally developed from data collected while the subject's hips and legs were constrained by the PSS. During free-dynamic lifting, the subject's pelvis was expected to rotate and change the length of muscles attached to the pelvis. The purpose of the model adjustment phase was, therefore, to determine the relative changes in muscle length due to pelvic motion.

Determination of muscle length consisted of a biomechanical analysis of the normalized EMG data collected from the subjects. This was accomplished by comparing the measured sagittal trunk moment from the force plate with the predicted sagittal trunk moment from the original muscle length-strength relationship for that gender (Granata and Marras, 1995a, Marras et al., 2001). Determination of the length-strength relationship included a systematic analysis procedure incorporating different inputs into the biomechanical model using the general form of equations 2.1a and 2.2 (Davis et al., 1998, Granata and Marras, 1993, Granata and Marras, 1995a, Marras and Granata, 1995, Marras and Granata, 1997b, Marras and Sommerich, 1991b, Marras and Sommerich, 1991a).

$$Force_j = Gain \times (EMG_t / EMG_{max}) \times Area_j \times f(Vel) \times f(Length) \qquad (Eq\ 2.1a)$$

Where:
$Force_j$ = the tensile force for muscle j.
Gain = the force producing capacity of the muscle per unit area (N/cm^2).
EMG_t = the integrated EMG value from the lifting exertion.
EMG_{max} = the maximum integrated EMG from the MVCs.
$Area_j$ = the maximum physiological cross-sectional area of muscle j.
$f(Vel)$ = the force-velocity modulation function.
Vel = Erector Spinae velocity as determined by the LMM.
$f(Length \times Adjust)$ = the length-strength modulation function.
Length = Erector Spinae muscle length as determined by the LMM.

Model gain was determined individually for each trial by iteratively comparing the predicted moment ($M_{x\text{-}pred}$) to the moment measured via the force plate (and translated to the subject's lumbo-sacral joint) and adjusting gain until the RMS error between the predicted and measured moments was minimized. This resulting gain was applied to all muscles. Gain was then multiplied by normalized EMG, (EMG_t / EMG_{max}). This result was multiplied by the muscle area, ($Area_j$), which is the maximum physiological cross-sectional area of muscle j as determined by Marras et al, (Jorgensen et al., 2001). Finally, this result was multiplied by outputs from the force-velocity, $f(Vel)$, and length-strength, $f(Length)$, modulation functions as quantified by Marras et al (Jorgensen et al., 2001).

As the purpose of the model adjustment phase was to determine an adjustment for muscle length, Equation 2.1a was modified to include this adjustment factor.

$$Force_j = Gain \times (EMG_t / EMG_{max}) \times Area_j \times f(Vel) \times f(Length \times Adjust) \quad (Eq\ 2.1b)$$

Where:
Adjust = the adjustment factor for muscle length.

The Adjust factor is the adjustment to erector spinae muscle length quantified by the results of the model adjustment phase. In the biomechanical model, the erector spinae, the internal oblique and the latissimus dorsi muscles act as agonists during sagittally symmetric lifting. Preliminary analyses adjusting the length of the internal oblique and latissimus dorsi muscles did not significantly affect the model. Therefore, a simplifying assumption was made that the erector spinae group are the primary muscles significantly altered by pelvic rotation during sagittally symmetric lifting exertions. It would be expected that antagonist muscle length would also be affected by changes in pelvic orientation. However, antagonistic muscle activity has been shown to be minimal during similar motions in other studies (Davis et al., 1998, Granata and Marras, 1995a) so an additional simplifying assumption was that changes in antagonist muscle force due to changes in pelvic orientation would be negligible.

The Adjust factor was incrementally increased from 1.00 to 1.03 in steps of 0.005 (changing muscle length by 0.5% each time) and the changes in gain, R^2 and the AAE between predicted and measured moments were observed. Initially, a wider range was used for the Adjust factor however when Adjust moved above 1. 30, gain rapidly moved outside of the physiologically valid range of 30 and 90 N/cm^2 (McGill et al., 1988, Reid and Costigan, 1987, Weis-Fogh, 1977) with a correspondingly rapid increase in AAE. Adjust values less than one were not used because the pelvis and back did not orient in such a way that would shorten the muscle. The value that produced the best combination of these variables was identified and used to adjust the model. Equation 2.2 then used the results of Equation 2.1b to predict the muscle moment imposed on the spine, ($M_{x\text{-pred}}$).

$$M_{x\text{-pred}} = \Sigma(r_j \times Force_j) \qquad (Eq\ 2.2)$$

Where:
$M_{x\text{-pred}}$ = the predicted muscle moment imposed on the spine.
r_j = the moment arm for muscle j.

The forces ($Force_j$ from Equation 2.1b) were multiplied by the moment-arm for that muscle, r_j and the resulting moments summed to determine $M_{x\text{-pred}}$.

Part Three: Testing the Validity of the Adjusted Model

The unadjusted model has been previously validated for straight-legged lifting and the goal of this validation effort was to achieve similar, or improved results while relaxing some of the constraints from the previous validation (Granata and Marras, 1995a). During the previous validation, subjects lifted loads of 18.2 and 36.4 kg at isokinetic angular velocities (0, 30, 60 and 90 deg/s) as well as free-dynamically (slow, medium and fast lift rates). While standing in an upright posture, the legs and pelvis of each subject were fastened to a rigid structure extending from the force plate. Isokinetic lifts were performed from a flexed trunk position of 45° to an erect posture at a rate subjectively controlled from video feedback. To perform free-dynamic exertions subjects were instructed to complete the entire motion in 2 s (free-dynamic slow), 1 s (free-dynamic medium), or as quickly as possible without jerking (free-dynamic fast). Averaged over all dynamic exertions, subject gains averaged 64.9 ± 27.6 N/cm^2, R^2 averaged 0.81 and AAE averaged 17.5 Nm (less than 15% of peak lifting moment).

The goal for validating the adjusted model was to achieve similar, or improved values for gain, R^2 and AAE compared to the previous model validation while allowing the subject to lift in a whole body, free-dynamic manner. Additionally, the adjusted model must also improve on those same model performance variables over the unadjusted model applied to the same data. Descriptive statistics describing the mean and variability of the model performance parameters were first performed. Analysis of Variance (ANOVA) procedures were used to test the significance of the model performance parameters (i.e., R^2, gain, and AAE) as a function of the independent variables.

RESULTS

As previously indicated, the aims of this study were achieved through a three-part process. In the instrument development phase, instrumentation was developed for the adjustment and validation of model performance during whole body free-dynamic lifting. In model adjustment phase, this instrumentation was used in an experiment designed to adjust the biomechanical model for whole body free-dynamic lifting. The model validation phase validated the adjusted model under whole body free-dynamic lifting conditions.

Part One: Development of Instrumentation to Allow for Whole Body Free-Dynamic Lifting in the Model

Goniometers were developed to achieve two objectives. First, the LSPM and PAM were developed to translate forces and moments measured at the force plate to the subject's lumbo-sacral joint and then rotate the translated forces and moments to align with the angle of the subject's pelvis during lifting. The LSPM provided data needed for translation of the forceplate measurements to the subject's lumbo-sacral level instantaneously throughout the lift. Once the measured moments had been translated to the subject's lumbo-sacral joint, they were instantaneously rotated by the PAM to account for pelvic orientation and pelvic motion during lifting.

Second, the Hip and Knee Monitors were developed to provide kinematic data on lower extremity positions. The Hip Monitor measured the position of the upper leg relative to the pelvis in the flexion-extension and abduction-adduction planes of motion. This data was used to determine possible changes in muscle length due to hip flexion. To provide an additional measure of lower extremity kinematics, the Knee Monitor was used to measure knee flexion during lifting.

L5/S1 Position Monitor

Construction of the L5/S1 Position Monitor

The LSPM consisted of a system of three potentiometers located at a fixed location 80 cm behind and 80 cm above the center of the force plate. A schematic of the LSPM is depicted in Figure 3.1. As shown in the schematic, the distance cable runs from a spool attached to potentiometer 1, through a guide and attaches to the subject at the lumbo-sacral level. As the spool turns to let out the distance cable, potentiometer 1 turns and the linear distance of the subject's lumbo-sacral joint is continuously reported. Similarly, potentiometers 2 and 3 attach to the distance cable guide. The angles the distance guide makes with the X and Z axes are instantaneously reported.

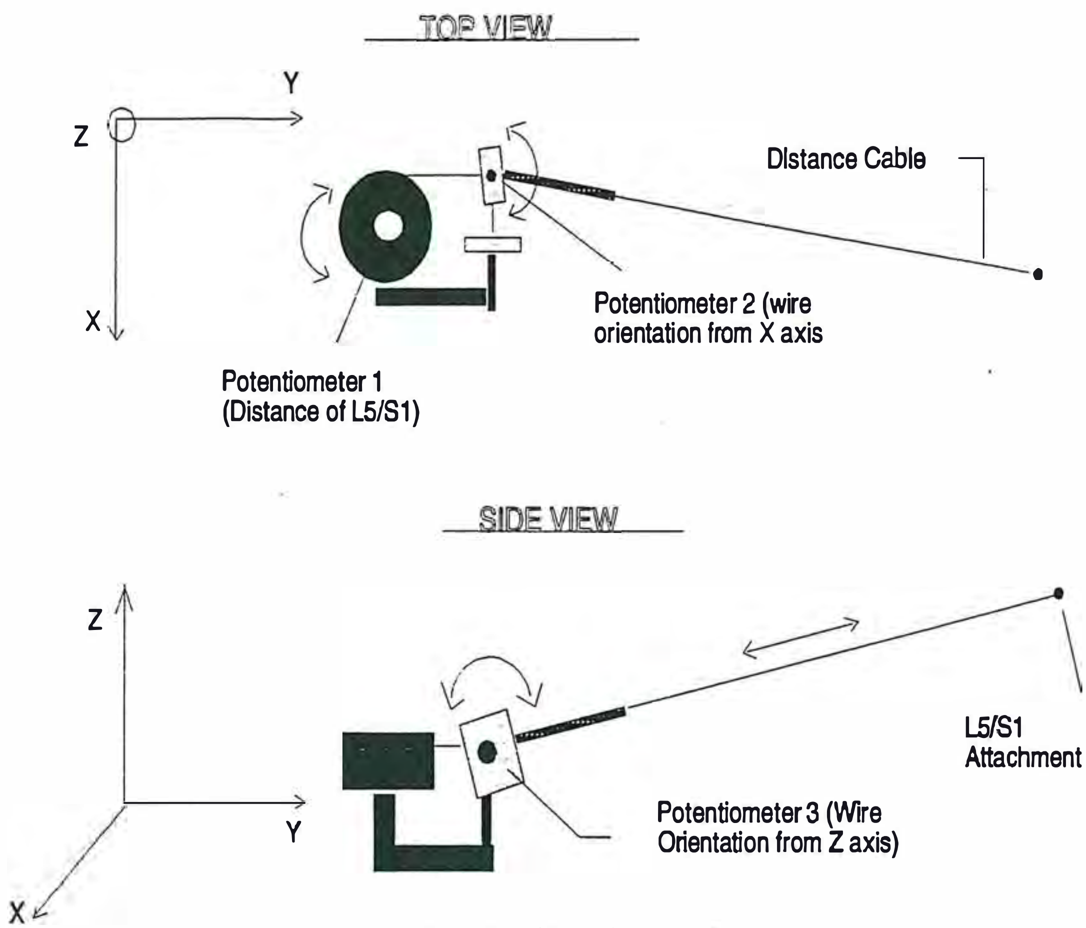

Figure 3.1: Schematic of the L5/S1 Position Monitor

Calibration of the LSPM

The LSPM was calibrated for each potentiometer (length, vertical angle and horizontal angle) independently. Length was calibrated by extending the distance cable to known lengths between 10 cm and 85 cm in 5 cm increments, collecting the resulting potentiometer voltage and using this data to create a regression equation for length. Vertical and horizontal angles were independently calibrated by moving the distance cable guide to known angles in 5° increments. The LSPM was calibrated vertically between 60° and 120° where 0° was defined as straight upward. The LSPM was calibrated horizontally between 40° counterclockwise and 140° where 0° was defined as directly left of the subject. Regression equations were then created for vertical and horizontal angles. R^2 values for all regression equations were greater than 0.98. Calibration data is shown in Table 3.1.

The LSPM calibrations were verified by using the LSPM to locate known positions in space within a region above the force plate that the subject might be expected to occupy and comparing the L5/S1 Monitor's reported locations with the known target locations.

Table 3.1: Calibration for the LSPM. (Regression equations shown at the bottom of each column)

Distance Cable Length (ρ)		Vertical Angle (σ)		Horizontal Angle (ϕ)	
Length (cm)	Volts	Angle (°)	Volts	Angle (°)	Volts
10	0.453	120	3.288	140	4.106
15	0.686	115	3.374	135	4.022
20	0.920	110	3.441	130	3.940
25	1.160	105	3.531	125	3.878
30	1.410	100	3.608	120	3.790
35	1.650	95	3.754	115	3.729
40	1.884	90	3.826	110	3.637
45	2.143	85	3.906	105	3.561
50	2.365	80	3.962	100	3.485
55	2.625	75	4.022	95	3.414
65	2.861	70	4.088	90	3.324
70	3.134	65	4.147	85	3.254
75	3.377	60	4.202	80	3.163
80	3.645			75	3.088
85	3.880			70	3.003
				65	2.933
				60	2.858
				55	2.767
				50	2.680
				45	2.646
				40	2.565
$y = 20.314x + 1.3694$ $R^2 = 0.9998$		$y = -63.064x + 328.43$ $R^2 = 0.9889$		$y = 64.093x - 17.281$ $R^2 = 0.9994$	

Pelvic Angle Monitor

Construction of the Pelvic Angle Monitor

The PAM consists of a system of two potentiometers attached to the LMM over the subject's lumbo-sacral joint. The potentiometers measure pelvic orientation by monitoring the angles of a guide to which they are connected (Figure 3.2). The distance cable from the LSPM connects to the guide on the PAM.

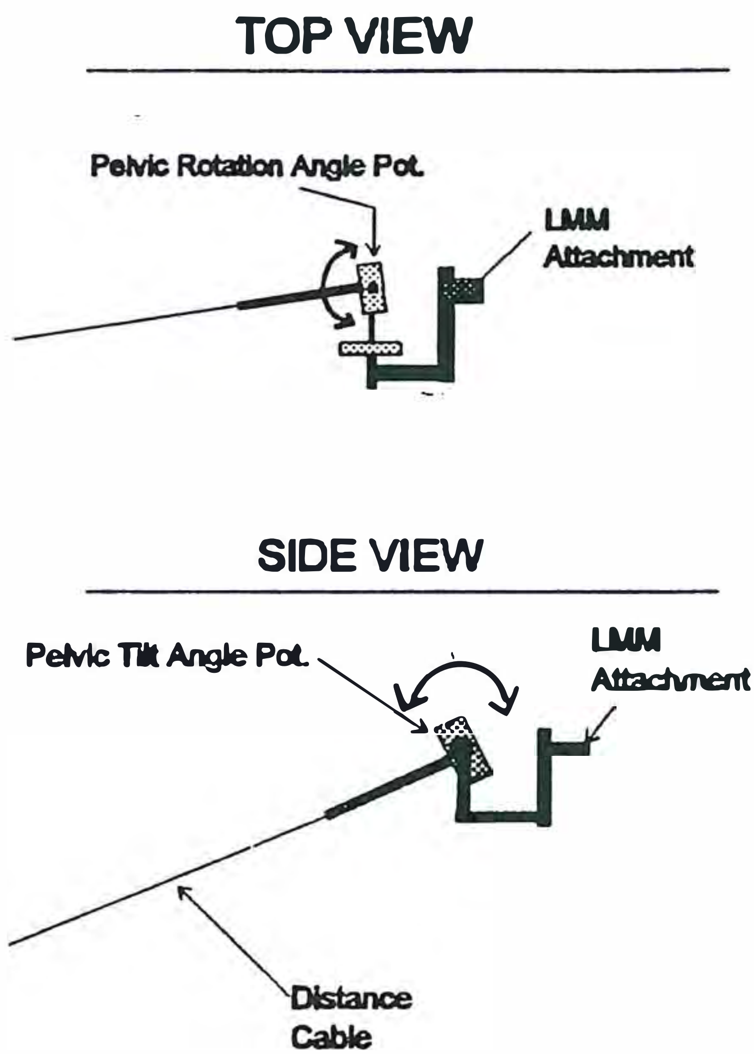

Figure 3.2: Schematic of the PAM

Calibration of the Pelvic Angle Monitor

The PAM was calibrated using a method similar to that employed with the LSPM. Vertical and horizontal angles were independently calibrated by moving the guide to known angles in 5° increments. The PAM was calibrated vertically between -45° and 45° where 0° was defined as perfectly horizontal and angles increased as the guide moved upwards. The PAM was calibrated horizontally between -90° and 90° where 0° was defined as perpendicular to the standing subject and angles increased to the subject's right. Regression equations were then created for vertical and horizontal angle. R^2 values for these regression equations were greater than 0.99. Calibration data is shown in Table 3.2.

Table 3.2: Calibration for the PAM. (Regression equations shown at the bottom of each column)

Vertical Angle		Horizontal Angle	
Angle (°)	Volts	Angle (°)	Volts
		-90	0.281
		-75	0.501
		-60	0.722
-45	3.996	-45	0.951
-30	3.765	-30	1.204
-15	3.510	-15	1.382
0	3.298	0	1.577
15	3.073	15	1.800
30	2.854	30	1.940
45	2.616	45	2.218
		60	2.435
		75	2.651
		90	2.908
$y = -65.607x + 216.62$ $R^2 = 0.9996$		$y = 69.819x - 110.47$ $R^2 = 0.9988$	

The PAM calibration was verified by placing the guide at known combinations of vertical and horizontal angles and comparing the measured angles with known values.

Mathematical Translation and Rotation of Force Plate Outputs by the L5/S1 Position Monitor and Pelvic Angle Monitor

The combination of L5/S1 Position and PAM was used to translate forces and moments measured at the forceplate to the subject's lumbo-sacral joint. The top portion of Figure 3.2 shows a schematic of how both monitors attach to a subject standing on the force plate. The bottom portion of Figure 3.3 is a representation of the force and distance vectors required to calculate moments at the lumbo-sacral joint.

An in-depth review of the mathematics behind this calculation is presented by Fathallah (Fathallah, 1995) who expands on the method employed by Granata and Marras (Granata and Marras, 1995a). A summary of this review follows: The moment vector measured at the force plate can be written as:

$$\vec{M}_{FP} = \vec{R}_{FP} \times \vec{F}, \qquad (1)$$

Where,

$\vec{M}_{FP}$ = The Moment vector measured at the force plate in three-dimensions (x, y, and z directions):

$$M_{FPx} = F_z \times R_{FPy} - F_y \times R_{FPz}, \qquad (2)$$

$$M_{FPy} = F_x \times R_{FPz} - F_z \times R_{FPx}, \qquad (3)$$

$$M_{FPz} = F_y \times R_{FPy} - F_x \times R_{FPy}, \qquad (4)$$

$\vec{F}$ = The applied external force vector, with components F_x, F_y and F_z in the x, y and z directions.

$\vec{R}_{FP}$ = The distance vector between the force plate and force vector $\vec{F}$ with three components R_{FPx}, R_{FPy} and R_{FPz} in the x, y and z directions.

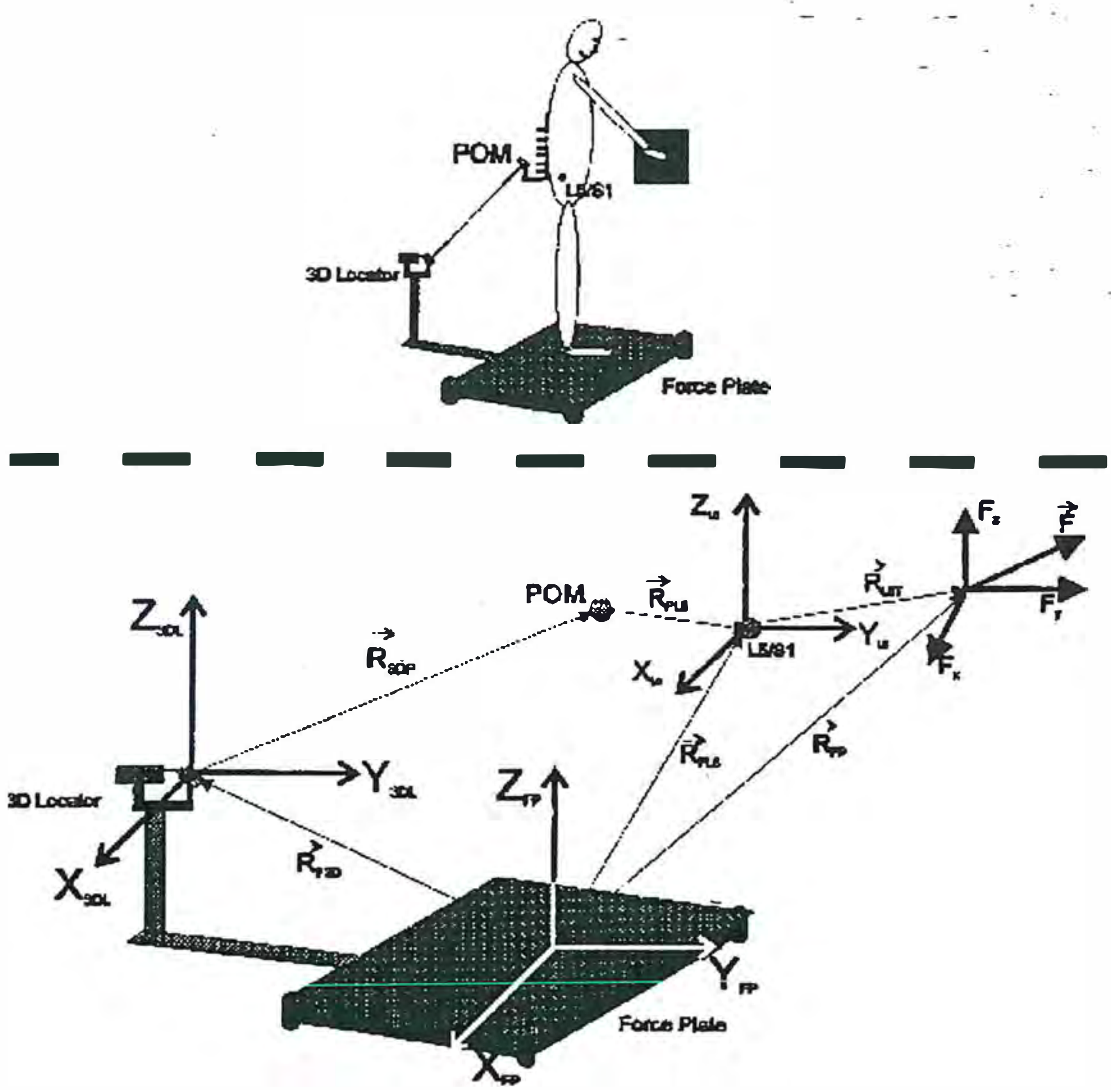

Figure 3.3. A 3-D representation of the vector translation from the force plate moment to about L5/S1.

The moment vector measured about L5/S1, $\vec{M}_{L5/S1}$:

$$\vec{M}_{L5/S1} = \vec{R}_{L5T} \times \vec{F}, \qquad \textbf{(5)}$$

Where,

$\vec{R}_{L5T}$ = The distance between L5/S1 and the force $\vec{F}$

From vector geometry (see the bottom half of Figure 3.3):

$$\vec{R}_{FL5} = \vec{R}_{F3D} + \vec{R}_{F3DP} + \vec{R}_{PL5}, \qquad \textbf{(6)}$$

And

$$\vec{R}_{FP} = \vec{R}_{FL5} + \vec{R}_{L5T}, \qquad \textbf{(7)}$$

Where,

$\vec{R}_{FL5}$ = The distance vector between the force plate and L5/S1, with three components R_{FL5x}, R_{FL5y} and R_{FL5z} in the x, y and z directions.

$\vec{R}_{F3D}$ = The distance vector between the force plate and the LSPM, with three components R_{F3Dx}, R_{F3Dy} and R_{F3Dz} in the x, y and z directions.

$\vec{R}_{3DP}$ = The distance vector between the LSPM and the PAM, with three components R_{3DPx}, R_{3DPy} and R_{3DPz} in the x, y and z directions.

$\vec{R}_{PL5}$ = The distance vector between the PAM and the subject's L5/S1 joint, with three components R_{PL5x}, R_{PL5y} and R_{PL5z} in the x, y and z directions. This is a known vector computed from subject anthropometry.

Hence:

Combining **(2)** and **(7)**:

$$M_{FPx} = F_z \times (R_{L5Ty} + R_{FL5y}) - F_y \times (R_{L5Tz} + R_{FL5z}), \qquad \textbf{(8)}$$

Expanding **(5)** and **(8)**:

$$M_{L5/S1x} = M_{FPx} + F_y \times R_{FL5z} - F_z \times R_{FL5y}, \qquad \textbf{(9)}$$

Similarly:

Combining (3) and (7):

$$M_{FPy} = F_x \times (R_{L5Tz} + R_{FL5z}) - F_z \times (R_{L5Tx} + R_{FL5x}), \quad \textbf{(10)}$$

Expanding (5) and (10):

$$M_{L5/S1y} = M_{FPy} + F_z \times R_{FL5x} - F_x \times R_{FL5z}, \quad \textbf{(11)}$$

Similarly:

Combining (4) and (7):

$$M_{FPz} = F_y \times (R_{L5Tx} + R_{FL5x}) - F_x \times (R_{L5Ty} + R_{FL5y}), \quad \textbf{(12)}$$

Expanding (5) and (12):

$$M_{L5/S1z} = M_{FPz} + F_x \times R_{FL5y} - F_y \times R_{FL5x}, \quad \textbf{(13)}$$

The force plate measures $\vec{F}$ and $\vec{M}_{FP}$ directly. The LSPM and the PAM provide $\vec{R}_{FL5}$. Thus, the moment about the L5/S1 joint, $\vec{M}_{L5/S1}$, is computed. The force vector $\vec{F}$ is also translated to the L5/S1 joint to provide a measure of the forces imposed on the L5/S1 joint.

One detail inherent in this calculation but not previously discussed should also be noted. The force exerted due to subject weight is initially removed from $\vec{F}$ by the force plate by "zeroing" the force plate. The weight of the upper body (above L5/S1) is then calculated from anthropometry (Dempster, 1955, Clauser et al., 1969) and added back in to the calculation of the external forces and the resulting force on the L5/S1 joint, $\vec{F}_{L5/S1}$ becomes:

$$F_{L5/S1x} = F_{xtfp}, \quad \textbf{(14)}$$

$$F_{L5/S1y} = F_{ytfp}, \quad \textbf{(15)}$$

$$F_{L5/S1z} = F_{ztfp} + F_{zbw}, \quad \textbf{(16)}$$

Where,

$F_{L5/S1x}$, $F_{L5/S1y}$, $F_{L5/S1z}$	= Forces at the L5/S1 joint in the x, y, and z directions.
F_{xtp}, F_{ytp}, F_{ztp}	= Measured forces translated from the force plate in the x, y, and z directions.
F_{zbw}	= Upper body weight (above L5/S1 level).

Hip Monitor

Construction of the Hip Monitor

The Hip Monitor consisted of a rod running alongside the upper leg and connected to the pelvis at a universal joint monitored by two potentiometers. The point of rotation of the universal joint was aligned with the hip joint center of rotation. One potentiometer measured flexion-extension motion while the other measured abduction-adduction of the hip. A schematic of the Hip Monitor is shown in Figure 3.4.

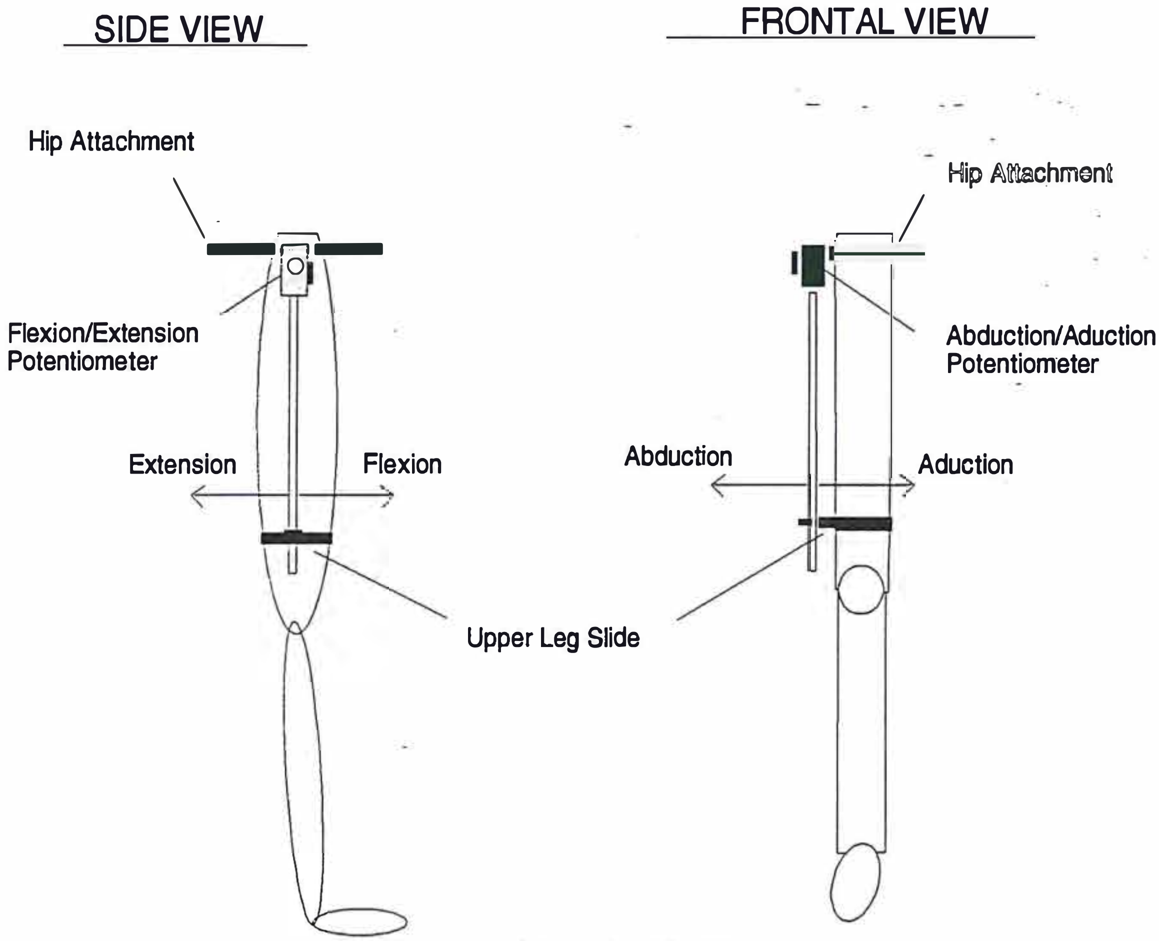

Figure 3.4. Schematic of the Hip Monitor.

Calibration of the Hip Monitor

The Hip Monitor was calibrated in a three-dimensional reference frame also used to calibrate the LMM. Sagittal flexion and ab/adduction angles were independently calibrated by moving the guide to known angles in 15° increments. The hip monitor was calibrated between -90° and 90° in each plane of motion where 0° was defined as straight down (to corresponds to an idealized subject standing perfectly upright). Sagittal angle increased with increasing hip flexion and ab/adduction angle increased with hip abduction. R^2 values for these regression equations were greater than 0.999. Calibration data is shown in Table 3.3.

Table 3.3: Calibration for the Hip Monitor. Regression equations shown at the bottom of each column.

Flexion Angle		Abduction Angle	
Angle (°)	Volts	Angle (°)	Volts
-90	1.52	-90	4.50
-75	1.72	-75	4.23
-60	1.95	-60	4.00
-45	2.18	-45	3.81
-30	2.42	-30	3.56
-15	2.63	-15	3.32
0	2.92	0	3.11
15	3.11	15	2.82
30	3.37	30	2.55
45	3.64	45	2.33
60	4.26	60	2.08
75	1.52	75	1.90
90	1.72	90	1.66
$y = 64.365x - 186.17$ $R^2 = 0.9991$		$y = -62.951x + 193.07$ $R^2 = 0.9991$	

The Hip Monitor calibration was verified by placing the leg rod at known combinations flexion and abduction angles and comparing the measured angles with the known values.

Knee Monitor

Construction of the Knee Monitor

The Knee Monitor consists of two strips of metal joined together by a potentiometer. The strips were affixed to the subject's upper and lower leg, with the joining potentiometer located over the knee joint. As the subject's knee bent, the metal strips rotated the potentiometer causing a change in voltage, which was then mathematically translated into knee flexion angles. In essence, the knee monitor acted as a hinge located at the knee joint. The angle of the hinge was instantaneously reported as the subject's knee flexed/extended. The design of the Knee Monitor has previously been employed to monitor joint motion in the upper extremity (Marras and Schoenmarklin, 1993, Schoenmarklin and Marras, 1993). A schematic of the knee monitor is shown in Figure 3.5.

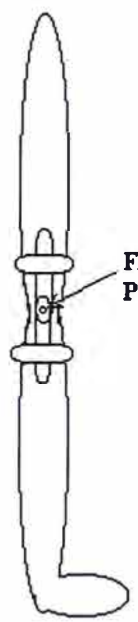

Figure 3.5: Schematic of the Knee Monitor.

Calibration of the Knee Monitor

The Knee Monitor was calibrated by positioning the metal strips at known angles between -90° and 90° where 0° was defined as straight (corresponding to a straight leg) and angles increased as the knee flexed. A regression equation was then created from the results for knee flexion. The R^2 value for this regression equation was greater than 0.999. Calibration data for the Knee Monitor is shown in Table 3.4.

Table 3.4: Calibration for the Knee Monitor. Regression equations shown at the bottom of each column.

Flexion Angle	
Angle (°)	Volts
-90	1.05
-75	1.26
-60	1.47
-45	1.72
-30	1.96
-15	2.19
0	2.41
15	2.65
30	2.87
45	3.09
60	3.30
75	3.53
90	3.75
$y = 66.151x - 159.02$ $R^2 = 0.9998$	

This Knee Monitor calibration was verified by placing the Knee Monitor in known positions and comparing the measured with known angles.

Part Two: Adjusting the model for whole body free-dynamic lifting exertions.

It was expected that the length-strength and force-velocity relationships of the trunk muscles attached to the pelvis would be influenced by hip flexion during the lift. Specifically, hip flexion was expected to affect the rectus abdominus, internal obliques, external obliques and erector spinae muscle pairs. Thus, this part of the research focused on determining what adjustments needed to be made to the model length-strength and force-velocity relationships to adjust the model for whole body free-dynamic lifting, if any adjustments were necessary.

Following the procedures previously described in the methods section of this report, the force-velocity and length-strength relationships were systematically adjusted and the effects of these adjustments on model performance were evaluated. Initially, adjusting the force-velocity did not result in an appreciable improvement to model performance while adjustments to the length-strength relationship yielded positive results. Thus analysis focused on adjusting the length-strength relationship.

The results of our analyses showed that changes to the length of the rectus abdominus, external and internal obliques did not result in improved model performance so analyses then focused on systematic adjustment of erector spinae muscle length.

Erector spinae length was systematically adjusted by applying a multiplier to muscle length. Multipliers of 1.000 (unadjusted), through 1.030 were applied in increments of 0.005 and their effects on model performance compared. Figures 3.6-3.8 show these effects.

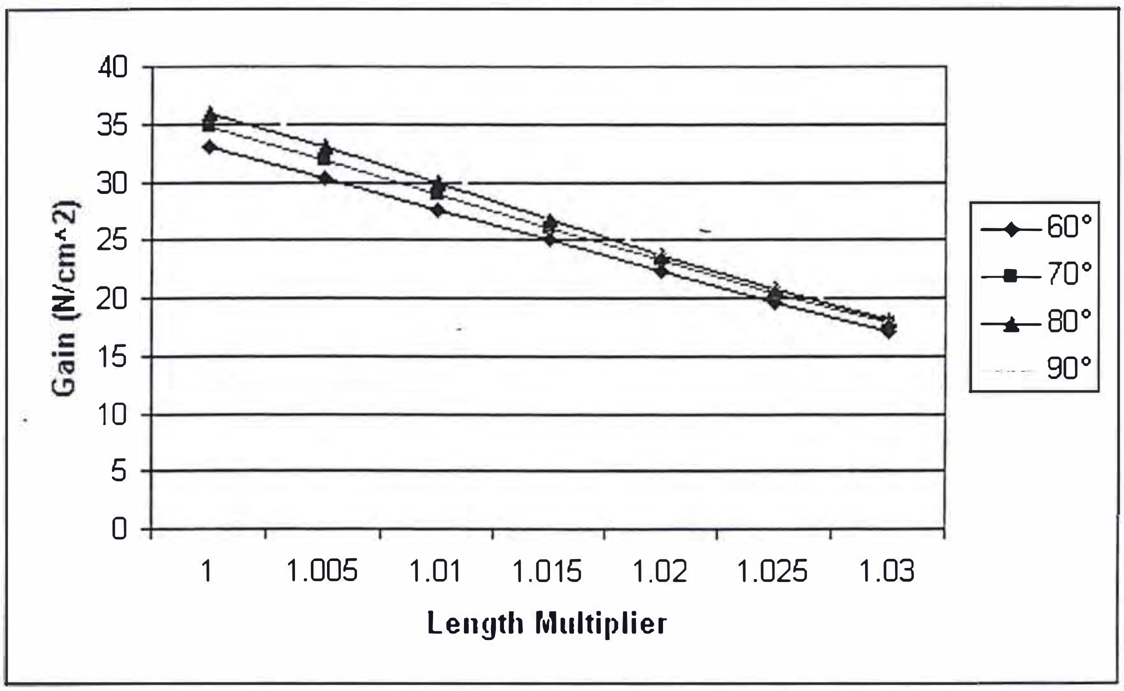

Figure 3.6: Average gain values versus erector spinal length multiplier by knee angle

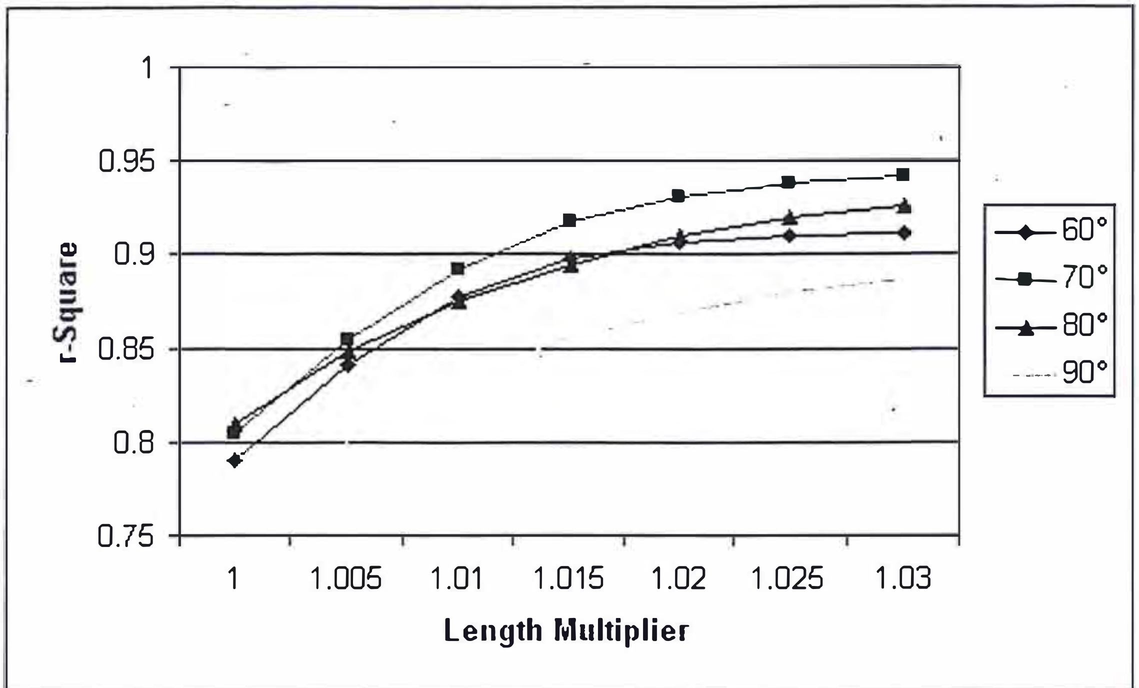

Figure 3.7: Average R^2 values versus erector spinal length multiplier by knee angle

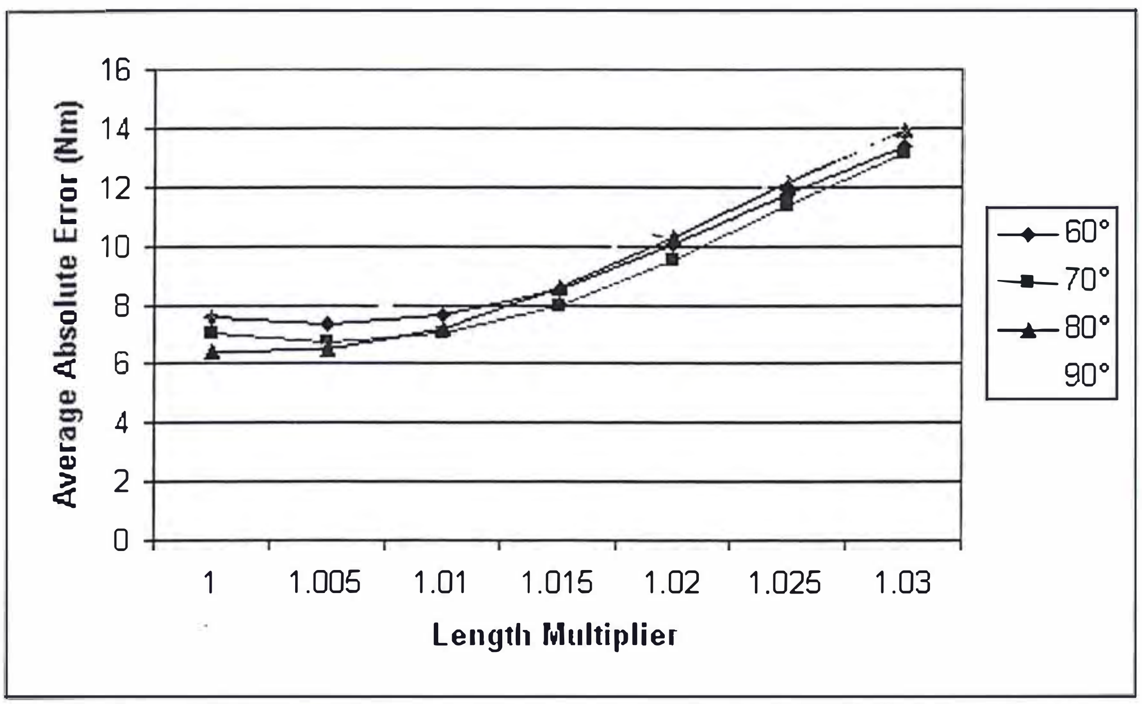

Figure 3.8: Average AAE values versus erector spinal length multiplier by knee angle

Gain consistently decreased with increasing erector spinae muscle length and R^2 increased with increasing erector spinae muscle length. Additionally, AAE decreased slightly between length multipliers 1.00 and 1.005 and increased thereafter. 1.005 was chosen as the multiplier because 1.005 caused a beneficial change in all model performance variables as well as minimizing AAE. Because the effect of the length multiplier on model performance variables followed the same trend for all knee angles, the multiplier 1.005 was applied throughout the entire lift.

Part Three: Validation of the adjusted model

The Analysis of Variance on the performance parameters of the original and adjusted models indicated that the adjusted model varied significantly from the original in predicting muscle gains and in the R^2 values between measured sagittal moment and moment predicted by the model (Table 3.5). Muscle gain was also significantly affected by the Model*Gender interaction.

Table 3.5: Analysis of Variance p-values for biomechanical model performance parameters (Significant values in bold).

	Gain	R^2	AAE
Model	**0.0001**	**0.04**	0.41
Model * Gender	**0.003**	0.61	0.41
Model * Height	0.88	0.90	0.74
Model * Weight	0.74	0.94	0.97
Model * Frequency	0.82	0.98	0.94
Model * Style	0.79	0.97	0.90
Model * Asymmetry	0.83	0.96	0.86

Significant Effects of Adjusted Model

The adjusted model produced lower gains than the original with model acting both as a main affect and as part of the Model*Gender interaction (see Tables 3.6 and 3.7). Overall, the adjusted model decreased predicted muscle gain by 9.0% which made this value more realistic. The adjusted model decreased model gain by 10% for males and 7.7% for females.

Thus, there was more of an effect of the adjustment for males. The adjusted model increased R^2 by 0.5% for both genders over the original model.

Table 3.6: Averages of model performance variables by model type. (Significant values in bold). (Standard deviations in parenthesis)

	Gain (N/cm^2)	R^2	AAE
Adjusted	**35.0** **(19.0)**	**0.843** **(0.193)**	15.1 (10.6)
Original	**38.4** **(20.6)**	**0.839** **(0.194)**	15.2 (10.6)

Effect of Adjusted Model on Model Performance Variables (Gain, R^2, AAE)

Summary statistics of model performance for the original and adjusted models are presented in Table 3.7. Overall, the adjusted model performed well, resulting in decreased gain and increased R^2 for all levels of all dependant variables (gender, height, weight, lifts/minute, lifting style, asymmetry). Similarly, AAE decreased for all levels of all dependant variables except for females and for the 13.6 kg load conditions where the original and adjusted models produced identical AAEs.

The adjusted model was relatively robust in the face of variations in experimental task and gender. Average gains for each dependant variable ranged from 30.0 N/m^2 to 39.9 N/m^2. R^2 values for each dependant variable ranged from 0.806 to 0.879 and AAEs ranged from 11.0 to 14.0 percent of measured sagittal moment. Starting height had the greatest effect on model performance. Gains decreased by 9.9 N/cm^2 for the ankle start height versus the knee start height. R^2 values decreased by 0.027 and AAE increased by 2.6 Nm on average. Gender had the second greatest effect. Males' gains were 5.8 N/cm^2 greater than females while male R^2's were 0.052 greater and AAEs that were 2.6 Nm greater than females on average. Squat lifting resulted in a 4.2 N/cm^2 increase in gain, a 0.002 decrease in R^2 and a 1.9 Nm decrease in AAE compared to stoop lifting. In asymmetry, sagittally symmetric lifts had 3.8 N/cm^2 lower gains, 0.072 higher R^2's and 1.1 Nm lower AAEs than lifts performed at 45° of asymmetry. Increasing weight from 6.8 and 13.6 kg increased gains by 3.2 N/cm^2, increased R^2 by 0.073 and decreased AAE 3.2 Nm. Finally, the 2 lifts per minute lifting frequency resulted in an average increase in gains of 1.7 N/cm^2, a decrease in R^2 of 0.016 and a decrease in AAE of 0.6 Nm when compared to the 8 lifts per minute frequency.

Table 3.7: Model performance variables from the original model by dependant variable. Standard deviations in parenthesis

		Original Model			Adjusted Model		
		Gain (N/cm^2)	R^2	AAE (N)	Gain (N/cm^2)	R^2	AAE (N)
Gender	Male	42.1 (20.9)	0.864 (0.177)	16.7 (11.6)	37.9 (19.0)	0.869 (0.174)	16.5 (11.6)
	Female	34.8 (19.5)	0.815 (0.207)	13.5 (7.1)	32.1 (18.4)	0.817 (0.206)	13.5 (7.2)
Height	Ankle	33.3 (16.0)	0.826 (0.201)	16.4 (11.0)	30.0 (14.4)	0.831 (0.199)	16.3 (11.0)
	Knee	43.5 (23.1)	0.853 (0.185)	13.8 (8.2)	39.9 (21.4)	0.855 (0.186)	13.7 (8.2)
Weight (kg)	6.8	40.2 (22.0)	0.875 (0.170)	13.6 (10.3)	36.6 (20.3)	0.879 (0.167)	13.4 (10.1)
	13.6	36.7 (18.8)	0.804 (0.209)	16.6 (9.0)	33.4 (17.3)	0.806 (0.209)	16.6 (9.1)
Frequency Lifts/min.	2	39.4 (21.1)	0.832 (0.200)	14.8 (9.3)	35.9 (19.5)	0.835 (0.199)	14.7 (9.4)
	8	37.5 (19.9)	0.847 (0.187)	15.4 (10.2)	34.2 (18.4)	0.851 (0.185)	15.3 (10.2)
Style	Squat	40.8 (23.3)	0.839 (0.207)	16.0 (11.3)	37.1 (21.5)	0.842 (0.206)	15.9 (11.3)
	Stoop	36.1 (17.1)	0.840 (0.180)	14.2 (7.9)	32.9 (15.7)	0.844 (0.179)	14.0 (7.9)
Asymmetry	Sag Sym	40.6 (20.7)	0.875 (0.157)	14.6 (10.0)	36.9 (19.0)	0.879 (0.153)	14.4 (9.9)
	45° CW	36.3 (20.2)	0.804 (0.220)	15.6 (9.5)	33.1 (18.6)	0.807 (0.220)	15.5 (9.6)

Overall, average model performance variables for each dependant variable for the adjusted model were either the same as the original model or slightly improved, though only the previously mentioned factors were significant

DISCUSSION

Part One: Development of Instrumentation to Allow for Whole Body Free-Dynamic Lifting in the Model

Essentially, two "sets" of goniometers were developed for this experiment. The LSPM and the PAM formed one set, designed to allow accurate translation and orientation of force plate measurements to the subject's lumbo-sacral joint. The Hip and Knee Monitors were the second set, designed to provide kinematic information on the lower extremity to assist in model adjustment.

The goniometers developed for this experiment have successfully accomplished their intended purpose by providing the information necessary to adjust the model for whole body free-dynamic lifting exertions as seen by the good gain, R^2 and AAE values achieved in model validation phase. Therefore, the LSPM and the PAM allows us now to accurately model whole body free-dynamic lifting exertions.

Lifting is a whole body exertion involving contributions from the hips and lower extremity as well as the back (Splittstoesser et al., 2000). Currently, there is limited quantification on the effects of individual and workplace variables on the kinematics of the hips and lower extremity. Investigations into the effects of such variables on trunk kinematics have shown effects due to fatigue, injury, lift frequency, weight, origin and destination of load (Davis and Marras, 2000a, Fathallah et al., 1998, Ferguson et al., 2000, Sparto et al., 1997a, (Hagen et al., 1995, Sparto et al., 1997b). Yet few have investigated the lower extremity at the same time (Sparto et al., 1997a, Sparto et al., 1997b, Hagen et al., 1995). By providing a way to monitor hip and lower extremity kinematics during lifting, the goniometers developed for this study allow researchers to measure hip and lower extremity kinematics in addition to trunk kinematics during quantitative investigations of lifting.

There is also limited quantification of realistic lifting styles. Many researchers have investigated stooped lifting (straight legs) and squat lifting (bent knee) lifting (Dolan et al., 1994, Duplessis et al., 1998, Ekholm et al., 1982, Garg and Saxena, 1985, Garg and Saxena, 1979, Hagen and Harms-Ringdahl, 1994, Hagen et al., 1995, Kumar and Garand, 1992, Leskinen et al., 1983, Leskinen, 1985, Luepongsak et al., 1997, Lindbeck and Arborelius, 1991, Potvin et al., 1991, Toussaint et al., 1992, Welbergen et al., 1991). However, there is evidence that natural lifts are not performed as pure stoop or pure squat lifts (Splittstoesser et al., 2000). There is also evidence that lifting style changes over time (Sparto et al., 1997a). Yet, there has been limited quantification of actual lifting styles in realistic lifting situations (Burgess-Limerick and Abernethy, 1997, Burgess-Limerick et al., 1995, Marras and Granata, 1997a, Sparto et al., 1997a, Splittstoesser et al., 2000). These goniometers allow quantitative investigation of lifting styles under realistic lifting situations.

The effects of hip, pelvic and lower extremity kinematics on trunk kinematics can also be investigated now that a means exists to easily control these variables though biofeedback provided by the goniometers developed during this study. For instance, a subject could perform a fatiguing exertion while wearing the LMM, Hip Monitor and Knee Monitor simultaneously. From these goniometers, the relative contributions of the trunk, hips, and lower extremity to lifting could be quantified as the subject fatigues. Other investigations, such as changes in the relative contributions of the trunk, hips and lower extremity to lifting under a low ceiling or within a narrow space can also be investigated.

In addition, these goniometers allow less obvious investigations into commonly encountered nonlifting exertions such as pushing and pulling where subjects "throw their weight" against a load in order to gain enough momentum to move it. The Hip and Knee Monitors can be used in such situations to quantify hip and lower extremity contributions. The LSPM can be used to quantify the three dimensional motion of the body as subjects "throw their weight" against a load and aid modeling of pushing and pulling.

Part Two: Adjusting the model for whole body free-dynamic lifting exertions.

As previously noted, the original model has been under development for 18 years. Over that time, it has been the subject of many improvements including advances in EMG estimation techniques, extensive quantification of the length-strength and force-velocity relationships, incorporation of eccentric modulations and gender-specific muscle anatomies. These improvements have resulted in steady increases in the accuracy of estimation of model gain and agreement between predicted and measured moment experienced at the lumbo-sacral joint (R^2 and AAE). This process produced the original model, which has proven very robust and reliable within its original constraints of lifting (e.g. hips secured, legs straight and within 45° of trunk flexion). The results of the model validation phase (discussed later) show that the original model also performs well when the aforementioned constraints were removed. Given the original model's good overall performance, it is not surprising that fine adjustments to muscle length were required, rather than gross modifications to the underlying theory or experimental variables such as the force-velocity or length-strength relationships. Thus, the original model only required a means of locating the lumbo-sacral joint during lifting while simultaneously determining pelvic orientation (accomplished by the LSPM and PAM) and an adjustment to muscle length to be fully operational for modeling whole body free-dynamic lifting without constraints on lower extremity motion and amount of trunk flexion.

The results of this study show how sensitive the model is to changes in the length-strength relationship and how crucial this relationship is to model performance. Erector spinae muscle length was adjusted by increasing length by 0.5% (multiplying original muscle length by 1.005). As shown in Table 3.8 of the results section, this small modification to muscle length has significant effects on model performance. Thus, accurate determination of muscle length is vital to optimal model performance in whole body free-dynamic lifting.

This study has also shown that the trends in model performance variables for all knee angles followed the same pattern with changes in muscle length, with the minimum AAE occurring when muscle length was multiplied by 1.005. Therefore, a single adjustment to erector spinae muscle length was applied to all conditions. The need for only a single multiplier for all knee angles indicates that pelvic tilt has a similar effect on erector spinae muscle length throughout a wide range of lower extremity postures. This adjustment results in a model that is directly applicable to modeling efforts on industrial jobs. The range of postures examined in this study cover all postures observed during whole body free-dynamic lifting (Marras et al., 1995a, Marras et al., 1993, Splittstoesser et al., 2000).

Physiological gain is defined as the muscle producing capability of the muscle per unit area. Within the model, gain also incorporates the contribution of unaccounted for factors. At extreme trunk angles, passive tissues provide some restorative force that will not be reflected by muscle activity. Small unmonitored muscles in the trunk add some force to trunk extension that is not reflected by the activity of monitored muscles. Inaccuracies in the length-strength and force-velocity relationships will result in changes in model gain as well as increases in AAE. The effects of these factors within the constraints of the original model are relatively minor, as seen by the physiologically valid gains and low AAE values from the original model. Thus, it was important that the adjusted model not produce large increases in gain during whole body, free-dynamic lifting, as such increases would indicate that unaccounted for factors were resulting in increased error. The adjustment to muscle length identified in this study was chosen because it removes some of the previously unaccounted for error from within the model gain term without resulting in an increase in AAE.

The increase in R^2 indicated in figure 3.6 at length multiplier 1.005 provides additional support for choosing that multiplier. R^2 within the model is a measure of the agreement between changes in moment measured during lifting and changes in moment estimated at by the model. An increase in R^2 indicates that the model is reacting to changes in muscle force more accurately than previously. Thus, increases in R^2 resulting from modifications to muscle length indicate that the model length-strength relationship is producing more accurate values. These values are then used to produce more accurate estimates of muscle force. Because the model estimates muscle force instantaneously, this results in a more responsive estimate of muscle force and, thereby, more responsive estimates of loads and moments imposed on the lumbo-sacral joint.

Part Three: Validation of the adjusted model

The validation phase of this experiment was conducted for several reasons. The first reason was to provide an independent data set to validate the adjusted model. The experimental conditions in the model validation phase were intentionally chosen to represent a range of extremes found in industry (Marras et al., 1995a, Marras et al., 1993) while simultaneously placing as few constraints on the subjects as possible. This was done to encourage realistic lifting, as might be seen in industrial environments.

The second reason was to address limitations in the previous model. Previously, subjects were restricted to sagittal bends of less than 45° with straight legs. The model validation phase addressed these limitations by testing more complex lifting tasks, including extremes of trunk bending (low origin heights) as well as knee and hip flexion (squat lifting).

The results show a consistent improvement across all dependant variables for model gain and R^2 and either a decrease in AAE or the same amount of error attributable to the adjustment to erector spinae muscle length. Table 4.1 shows the "percent improvement" (decrease in gain, increase in R^2, decrease in AAE) in the adjusted model when compared to the original model. Gain improved 7% to 10% depending on the dependant variable in question with most improvements approximately 9%. At the same time, AAE either remained constant or was slightly reduced while R^2 increased consistently. The reduction of AAE indicates a more accurate estimation of spinal loads. This implies the adjustment to muscle length has fine-tuned the model's application of the length-strength relationship in estimating spinal loading, resulting in more realistic estimates of spinal loads.

Table 4.1: "Percent improvement" (decrease in gain, increase in R^2, decrease in AAE) of the adjusted model over the original model

		Gain	R^2	AAE
Gender	Male	10.00	0.46	1.20
	Female	7.47	0.25	0.00
Height	Ankle	9.91	0.61	0.60
	Knee	8.28	0.23	0.72
Weight (kg)	6.8	8.96	0.46	1.46
	13.6	8.74	0.37	0.00
Frequency Lifts/min.	2	8.88	0.36	0.67
	8	8.82	0.35	0.65
Style	Squat	9.09	0.36	0.00
	Stoop	8.86	0.48	1.41
Asymmetry	Sag Sym	8.89	0.34	1.37
	45° CW	8.82	0.25	0.00

The summary of model performance in Table 3.5 indicates the adjusted model performed well throughout a wide range of whole body free-dynamic lifting conditions based on the model performance variables. The majority of individual trials resulted in high R^2 values (82% and 71% of trials greater than $R^2 = 0.80$ for males and females, respectively) with physiologically valid muscle gains (gains around 35 N/cm^2) and low AAEs for the prediction of sagittal moment (for males AAE = 11.3% of sagittal moment, for females AAE = 14.3% of sagittal moment).

Table 4.2 shows the results of an analysis of variance on the model performance parameters for the adjusted model. The table shows a number of significant effects on the model performance parameters due to dependant variables. While these effects are significant at the $p<0.05$ level, their biomechanical effects are relatively small in magnitude. From Table 3.5, the greatest changes in R^2 and AAE due to different experimental conditions were 0.073 and 3.29 N/cm^2 respectively, which occurred when load lifted changed from 6.8 to 13.6 kg. Similarly, the greatest effect on gain due to different levels of any of the dependent variables was 9.9 N/cm^2 when comparing males to females. However, from Table 4.2, this effect was not found to be significant. The gain effect from the other dependant variables, though relatively small, may be due to a number of factors including differences attributable to the length-strength and force-velocity modulation factors. Thus, investigation of factors affecting gain remains an area where further research may be beneficial to efforts to model and understand whole body free-dynamic lifting.

Table 4.2: Analysis of Variance p-values for biomechanical model performance parameters of the adjusted model (Significant values in bold).

	Gain	R^2	AAE
Gender	0.23	**0.03**	0.13
Height	**0.0001**	0.06	**0.0001**
Weight	**0.0001**	**0.0001**	**0.0001**
Frequency	**0.001**	0.1	0.28
Style	**0.003**	0.72	**0.006**
Asymmetry	**0.01**	**0.0001**	0.076

Predicted Forces and Moments at L5/S1

Tables 4.3 and 4.4 show the forces and moments imposed at the lumbo-sacral joint as predicted by the original and adjusted models. On the whole, both models predicted similar average forces and moments. This similarity between the original and adjusted models provides some indication of the robustness of the original model. The basic theories behind the development of the original model outlined in the introduction produced a model that only needed minor modifications (system of goniometers to translate/orient force plate measurements to the lumbo-sacral joint, adjustment to erector spinae muscle length) to perform well for modeling whole body free-dynamic lifting.

Table 4.3: Predicted forces and moments from the adjusted model. Standard deviations in parenthesis

		Lateral Shear (N)	A-P Shear (N)	Compression (N)	Sagittal Moment (Nm)	Resultant Moment (Nm)
Gender	Male	143.8 (137.4)	683.5 (506.7)	4195.5 (1270.7)	161.8 (63.3)	213.2 (99.0)
	Female	315.6 (473.6)	462.6 (232.4)	3252.3 (1146.9)	113.0 (47.7)	157.2 (70.8)
Height	Ankle	274.3 (427.2)	691.6 (530.2)	4012.7 (1432.5)	146.2 (66.6)	196.0 (97.6)
	Knee	184.4 (266.5)	457.4 (185.8)	3443.4 (1084.9)	128.9 (53.8)	174.9 (81.5)
Weight (kg)	6.8	205.7 (334.5)	509.6 (330.1)	3394.5 (1208.8)	124.4 (57.0)	167.7 (82.9)
	13.6	252.6 (379.5)	638.4 (474.1)	4061.0 (1305.1)	150.8 (62.2)	203.1 (94.4)
Frequency Lifts/min.	2	228.8 (349.9)	567.7 (387.2)	3703.4 (1265.6)	136.9 (59.7)	185.2 (90.3)
	8	229.3 (366.7)	579.8 (437.6)	3748.9 (1335.2)	138.2 (62.6)	185.5 (90.8)
Style	Squat	235.9 (385.9)	623.3 (511.4)	3877.4 (1451.9)	144.7 (68.4)	196.4 (98.4)
	Stoop	223.3 (328.8)	524.8 (276.3)	3576.9 (1112.7)	130.5 (52.1)	174.4 (80.7)
Asymmetry	Sag Sym	194.9 (281.3)	585.6 (410.4)	3757.6 (1251.5)	140.1 (60.0)	177.4 (77.7)
	45° CW	263.7 (419.6)	561.6 (415.7)	3694.3 (1348.8)	134.9 (62.2)	193.4 (101.3)

Table 4.4: Predicted forces and moments from the original model. Standard deviations in parenthesis

		Lateral Shear (N)	A-P Shear (N)	Compression (N)	Sagittal Moment (Nm)	Resultant Moment (Nm)
Gender	Male	143.7 (138.2)	785.3 (521.6)	4237.2 (1286.4)	159.9 (62.6)	211.2 (99.3)
	Female	320.2 (480.0)	540.3 (239.3)	3289.6 (1286.4)	112.1 (47.7)	156.1 (70.1)
Height	Ankle	277.6 (433.6)	787.0 (544.7)	4058.2 (1451.5)	144.7 (65.8)	194.4 (98.1)
	Knee	186.0 (269.1)	541.5 (190.8)	3477.7 (1102.3)	127.6 (53.5)	173.3 (80.6)
Weight (kg)	6.8	207.5 (338.9)	595.9 (339.6)	3424.9 (1224.6)	122.8 (56.4)	165.9 (82.0)
	13.6	255.8 (384.8)	730.9 (487.9)	4110.7 (1323.5)	149.5 (61.5)	201.9 (94.6)
Frequency Lifts/min.	2	231.4 (354.8)	657.3 (398.9)	3742.4 (1284.8)	135.4 (59.1)	183.6 (90.7)
	8	231.6 (371.5)	669.7 (449.9)	3789.6 (1354.0)	136.8 (61.9)	184.0 (89.9)
Style	Squat	238.3 (391.1)	713.4 (524.6)	3914.1 (1468.5)	143.0 (67.5)	194.6 (98.8)
	Stoop	224.9 (333.4)	614.2 (287.5)	3619.7 (1136.1)	129.3 (51.8)	173.1 (79.6)
Asymmetry	Sag Sym	196.6 (284.3)	675.8 (422.8)	3794.8 (1267.4)	138.5 (59.3)	175.5 (76.6)
	45° CW	267.0 (425.9)	651.0 (427.4)	3736.8 (1371.0)	133.7 (61.6)	192.2 (101.7)

LIMITATIONS

There are a few limitations inherent in this research, which must be considered when assessing results of these studies. At this point, the model is only capable of assessing active trunk forces and passive spinal loading within the 45° of flexion. It is not sensitive to passive loading of the spine beyond 45°. It is possible that some MMH activities do involve extreme trunk flexion (greater than 45° of sagittal flexion) which would rely increasingly on passive structures of the low back. However, surveillance studies have demonstrated that trunk flexion in excess of 45° account for less than 5% of industrial MMH lifts (Marras et al., 1995a, Marras et al., 1993).

Additionally, the validation phase of this study investigated stoop (straight leg) lifting from ankle height, which required extreme trunk flexion from most subjects. The model performance variables were not severely affected by these exertions. Thus, neglecting passive spine loading does not present a large problem in the current study. Nevertheless, inclusion of passive spine loading in the model would produce more accurate model results. Research into methods to model passive spine loading during whole body, free-dynamic lifting is currently underway at The Biodynamics Laboratory.

The model currently models the spine as a straight rod, articulating at the lumbo-sacral joint. This limits the model to estimating spine loading at the lumbo-sacral joint and may affect the model's estimates of spine loading at the lumbo-sacral joint through inaccuracies in estimation of the angle of the L5 vertebral body with respect to the pelvis. Current research is underway to estimate individual intervertebral angles of the lumbar spine based on the overall trunk angle measured by the LMM.

The model currently estimates erector spinae moment arm using a single prediction based on an estimate derived from individual anthropometry. It has been shown that erector spinae muscle moment arms change during spinal motion (Macintosh et al., 1993). Research is underway to create an individual estimate of erector spinae moment arm based on anthropometry and intervertebral angles. This research can be combined with the previously mentioned research predicting intervertebral angles from LMM angle to add an individual, dynamic, estimate of erector spinae moment arm to the model.

The length-strength and force-velocity relationships were developed during previous studies from average values of a subject population (Marras et al., 2001). Individual differences in fiber type may affect these relationships causing an individual to have a different length-strength or force-velocity relationship from the global one used by the model. The same technique mentioned previously for determining gains for tasks where the subject cannot be confined to a force plate can also be used to create individual length-strength and force-velocity relationships. This will result in a more individually accurate model.

As previously noted, the relationships used in this model are based primarily on The Ohio State University graduate and undergraduate student population. Thus, the model is based on data collected from young adults. As people age, there may be changes to muscle orientation, the length-strength relationship or the force-velocity relationship. Therefore, it might be suggested that this population, while representative of young adults, may not fully represent older workforces such as those found in the automotive industry. Recent studies of workers in industry past college age resulted in values for gain, R^2, and AAE that were comparable to those found using student subjects in previous studies (Marras et al., 1997, Marras et al., 1999b). However, the research in these studies was not broken down by age. Therefore, further research is warranted to verify the model's applicability to older workers.

Workers with LBDs are often unable to perform the MVCs required by the modeling. As morbidity statistics indicate, a significant number of workers have LBDs. These injuries affect the kinematics and muscle activation patterns of the injured worker and would be expected to alter spinal loading (Burdorf et al., 1995, Ferguson et al., 1990, Marras et al., 1995b, Chen et al., 1998). Techniques to determine the MVC values of injured workers are currently under development.

The subject is currently still confined to the force plate during lifting. Researchers at The Biodynamics Lab are currently experimenting with implementing a technique whereby the subject performs standard lifts on the force plate at the beginning of the study. The subject then is disconnected from the LSPM and the PAM and steps off of the force plate. The study is then performed free-dynamically without constraining the subject to the force plate and the gains from the standard lifts are applied. This allows modeling of tasks that require the subject to move their feet. However, due to its size, the force plate itself is restricted to use in the laboratory.

The internal and external oblique muscle lengths were not adjusted during this study because adjustment of these muscles did not result in improvements to model performance parameters. This may be due to the sagittally symmetric nature of the experimental tasks performed during model adjustment phase of this study. Asymmetric tasks modeled during the model validation phase of this study indicate the model performs well under asymmetric lifting conditions. However the internal and external oblique muscle lengths may need adjustment during pure twisting exertions, where they are also active. Therefore, further study into the effects of twisting motions on internal and external oblique muscle length is warranted.

CONCLUSIONS

This study has successfully developed and validated a method for modeling spine loading during whole body, free-dynamic lifting. This method consists of a system of goniometers and an adjustment to the model's erector spinae muscle length. The adjusted model produced very good model performance parameters, including high R^2's between the predicted and measured moment, physiologically valid muscle gain values, and small magnitudes of error between the predicted and measured moment (AAE).

Development of the goniometers used in this study (LSPM, PAM, Hip Monitor and Knee Monitor) allowed accurate modeling of more realistic lifting exertions than was possible using the previous model. In addition, these goniometers will allow validation of future modifications to the model by providing instantaneous externally measured spinal loads for comparison during whole body free-dynamic lifting while simultaneously opening up new areas of exploration. These goniometers will make possible the modeling and quantification of lifting kinematics of the pelvis and lower extremity in addition to the trunk. In addition, more kinds of exertions may now be modeled, including whole body free-dynamic lifting and pushing and pulling exertions.

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FUTURE PUBLICATION

"The Effects of Box Location on Trunk, Hip and Knee Kinematics"