RELATIVE CONTRIBUTIONS OF THE LUMBAR SPINE AND PELVIS TO TRUNK MOTION DURING SAGITTAL PLANE MANUAL MATERIALS HANDLING

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ABSTRACT

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Eleven males and 11 females participated in a sagittal plane motion analysis study in which they lifted and lowered a symmetric load under two different conditions. Independant variables were *sex* (Male vs. Female), *lifting condition* (Freestyle vs. Constrained), and *direction* (Lift vs. Lower). The Constrained condition simulated handling loads in an industrial bin. The Freestyle condition had no obstruction. Dependant kinematic variables were mean peak trunk, pelvis, and knee flexion, and mean peak percent of maximum lumbar spine flexion ($\theta_{\%LUM}$). The sagittal plane pattern of movement between the lumbar spine and pelvis was studied by monitoring angular displacement time-histories, and via the calculation of phase angles between the two structures. Dependant kinetic variables were mean and peak thoracic and lumbar erector spinae EMG levels.

Synchronous movement was illustrated between the lumbar spine and pelvis for all conditions but the constrained lift. Sagittal plane movement between these two structures was more sequential for the constrained lift. No *sex* effect existed for any variables. $\theta_{\%LUM}$ was significantly greater during the lift than the lower, indicating slightly higher risk

for low back injury during lifting. Mean peak trunk flexion was substantially greater for the Constrained condition versus the Freestyle condition. This was a result of an increase in mean peak pelvic flexion. $\theta_{\%LUM}$ remained constant between these two conditions, and did not reach the level of flexion required to elicit the flexion-relaxation phenomenon.

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ABBREVIATIONS

EMG. Electromyography

ESA. Erector spinae aponeurosis

FRP. Flexion relaxation phenomenon

H. Horizontal distance between the load and ankle joint

IL-L. Iliocostalis Lumborum Pars Lumborum

IL-T. Iliocostalis Lumborum Pars Thoracis

L_i. An individual lumbar vertebra at the *i*th level (i=1,2,3,4,or 5)

LBD. Low back disorder

LBP. Low back pain

LES. Lumbar erector spinae muscle

MVC. Maximum Voluntary Contraction

TES. Thoracic erector spinae muscle

LT-L. Longissimus Thoracis Pars Lumborum

LT-T. Longissimus Thoracis Pars Thoracis

L_4/L_5. Intervertebral joint between the 4th and 5th lumbar vertebrae

V. Vertical distance between the load and ankle joint

$\theta_{\%LUM}$. Peak percent of maximum lumbar spine flexion

Chapter 1

INTRODUCTION

The study of lumbar spine mobility during lifting and lowering tasks is critical to the quantification of low back injury risk in the workplace. Epidemiologic studies show that billions of dollars are spent annually on the problem of low back pain (LBP), which is one of the most commonly cited problems for lost work time in industry and Workers Compensation claims (Frymoyer, 1988; Pope *et al.*, 1991; Chase, 1992). Manual materials handling tasks involving lifting of loads account for the majority of occupationally related risk of low back disorder (LBD) (Bigos *et al.*, 1986; Spengler *et al.*, 1986; Snook, 1989; Videman *et al.*, 1990). Most recently, Marras *et al.* (1993) were able to develop a multivariate vector in which lifting frequency, maximum sagittal trunk angle, and maximum load moment were included as 3 of 5 factors for the determination of high risk LBD situations.

It is well known that movement of the trunk in the sagittal plane is accomplished by flexion and extension of both the spine and the pelvis (Farfan, 1975; Gracovetsky *et al.*, 1977; Kippers and Parker, 1984; Mayer *et al.*, 1984; Anderson *et al.*, 1985; McGill and Norman, 1986; Potvin *et al.*, 1991; Paquet *et al.*, 1994; Ursulak and Potvin, 1994, 1995; Nelson *et al.*, 1995; Esola *et al.*, 1996; McClure *et al.*, 1997). There is, however, an ongoing debate as to whether the spine and pelvis are mobile at the same time (synchronous), or whether they are mobile at separate times (sequential). It must be noted that although the thoracolumbar spine is considered a single structure, sagittal motion originates primarily from the lumbar region since mobility in this plane generally decreases from the lower lumbar to the upper thoracic regions (Tanz, 1953; Allbrook, 1957; White and Panjabi, 1975; White and Panjabi, 1978). Thus, the magnitude of spinal flexion in the sagittal plane is mainly due to that of the lumbar spine. While the amount of lumbar spine flexion is controlled up to a certain degree by the activity of the erector spinae muscle group, extreme angles of flexion are supported by posterior spinal ligaments as the lumbar erector spinae become inactive (Allen, 1948; Floyd and Silver, 1951, 1955; Portnoy and Morin, 1956; Carlsoo, 1961; Pauly, 1966).

Primary passive resistance to lumbar spine flexion comes from (a) the interspinous and supraspinous ligaments (located on the dorsal spinal surface), (b) the capsular ligaments of the apophyseal joints, and (c) the intervertebral disc (Adams & Hutton, 1983b). In cases of hyperflexion, supraspinous and interspinous ligaments have been reported as being the first ligaments to sprain, followed then by the apophyseal capsular ligaments (Adams *et al.*, 1980). Repetitive loading of the lumbar spine in a flexed posture has also been observed to have degenerative effects on the annulus fibrosus of the intervertebral discs (Adams and Hutton, 1983a). If the supraspinous, interspinous and capsular ligaments are sprained, then increased stress is placed on the posterior annular fibres of the intervertebral disc during flexion. This leaves the disc succeptible to prolapse if subjected to high compressive forces (Adams & Hutton, 1983b).

Some researchers promote full lumbar flexion during lifting so that ligaments are preferentially used for extensor moment production. This is because they attribute the poor efficiency of the lower back musculature to result in higher compression and shear values at the intervertebral joint and can cause mechanical failure of the lumbar spine (Gracovetsky *et al.*, 1977). Conversely, Potvin and colleagues (1991) have suggested that maintaining a more lordotic lower back will allow the erector spinae muscles to dominate the production of extensor moment and, in turn, resist much of the anterior shear imposed on the lumbar spine intervertebral discs. Maintaining back extensor activity during spinal loading also provides for greater stability as the muscles can actively respond to perturbations.

MacDonald *et al.* (1997) completed one of the largest demographic studies on the occurrence and cost of both nonrecurrent and recurrent LBP claims to date. Data were collected over a 5 year and 2 month duration, throughout 44 states of the USA. With a study population of 110 983, it was found that males accounted for the majority (71%) of LBP claims while females accounted for a much smaller amount (MacDonald *et al.*, 1997). The mean nonrecurrent, recurrent, and total LBP claim costs were 20.7%, 6.6%, and 17.2% greater for male claimants as compared with female claimants (MacDonald *et al.*, 1997). Studies previous to this one have also shown that males have a higher incidence of LBP (Volinn *et al.*, 1991; Zwerling *et al.*, 1993; Butler *et al.*, 1995). Scientists know very little about why these differences exist. One may argue that they are due to the fact that males occupy a larger percentage of the industrial work force, yet this still does not explain why their injuries are more severe. Ursulak and Potvin (1995) have provided some preliminary evidence that males and females may perform sagittal plane lifts with different spine-pelvis kinematics. Results showed that males tended to perform these tasks with greater spine flexion, while females did so with greater pelvic flexion (Ursulak and Potvin, 1995). These kinematic differences may be linked to the risk of low back injury.

Purpose

The study of Ursulak and Potvin (1995) used a small sample size and was confounded with some lost data. Hence it was one of the goals of this study to replicate the aforementioned one to determine if their reported sex differences in spine - pelvic kinematics do actually exist. The data may provide some insight into the observed sex differences in injury statistics. It was also an objective of this study to use more quantitative methods than previously used in the determination of the pattern of sagittal plane movement between the lumbar spine and pelvis. Finally, studies have shown differences in peak lumbar spine and pelvic flexion angles when comparing squat to stoop techniques, yet none have analyzed these variables for comparisons between lifts and lowers within any type of manual materials handling technique.

Hypotheses

1. Loaded Flexion (lower) and extension (lift) of the trunk are accomplished via synchronous sagittal angular displacement of the lumbar spine and pelvis.

While these two segments may be flexing and extending at different rates from one another, at no point during trunk sagittal plane motion will one segment be in motion while the other is relatively stationary. Calculations of relative phase angles by Burgess-Limerick *et al.* (1992) have been the only to date to attempt to mathematically quantify the pattern of movement between these two segments.

2. During the lift and lower for both freestyle and constrained situations, males will have higher peak lumbar spine flexion angles than females. Conversely, females will achieve greater peak pelvic flexion angles than males.

Injury claim studies over the past 7 years have shown that males have a higher incidence of occupationally related low back disorders (Volinn *et al.*, 1991; Zwerling *et al.*, 1993; Butler *et al.*, 1995; MacDonald *et al.*, 1997). Ursulak and Potvin (1995) found during three variations of lifting that males had significantly greater peak lumbar spine flexion angles while females had significantly greater peak pelvic flexion angles. These findings provide some preliminary evidence that males have a higher risk of sustaining a low back injury during lifting, although their study was conducted with only five male and five female subjects.

3. The lowering task (loaded flexion) will be performed with greater peak flexion of the lumbar spine than the lifting task (loaded extension) for both male and female subjects.

Few experiments have been conducted to compare peak lumbar spine flexion between lifting and lowering of loads. The basis for this hypothesis lies behind the fact that spine forward flexion is a result of gravitational forces acting on the upper body, with the back muscles acting eccentrically to control the rate at which this occurs (Bogduk and Twomey, 1991; Oliver and Middleditch, 1991). Thus, an individual is actually being pulled down to the ground under the force of their upper body and load, when lowering a load. Therefore, it is hypothesized that the lumbar spine is likely to flex more at the end of lowering the load, as opposed to the beginning of lifting it.

4. Peak trunk flexion will be greater for both males and females during the constrained situation than the freestyle situation.

The constrained condition will decrease the amount of flexion permitted at the knee joint when performing the experimental tasks. Thus, in order to compensate for this restriction, an increase in trunk flexion will occur in order for the tasks to be completed. Upon comparing squat versus stoop lifting, substantial increases in trunk flexion have been observed due to the more extended knee posture during the stoop lift (Park and Chaffin, 1974; Garg and Herrin, 1979; Potvin *et al.*, 1991).

5. Increases in peak lumbar spine flexion will contribute to the greater trunk flexion, and will result in larger passive contributions to the extensor moment being generated (flexion- relaxation phenomenon). The flexion-relaxation phenomenon will be illustrated via decreased EMG amplitude of the lumbar erector spinae muscles at the lowest point of the lower, indicating a corresponding reduction in muscular activity.

Increases in lumbar spine flexion have been shown to contribute to the requirement of greater trunk flexion angles while handling loads (Potvin *et al.*, 1991; De Looze *et al.*, 1993; Ursulak and Potvin, 1994). The constrained condition in this study will cause an increase in the degree of lumbar spine flexion as compared to the freestyle condition. Consequently, this magnitude of flexion will be substantial enough to elicit the flexion-relaxation phenomenon. This event has been illustrated in previous studies by the silencing of the LES muscle group

(Allen, 1948; Floyd and Silver, 1951, 1955; Portnoy and Morin, 1956; Carlsoo, 1961; Pauly, 1966).

6. Although it is hypothesized that the constrained condition will cause an increase in peak lumbar spine flexion, the main contribution to the increase in peak trunk flexion will come from forward rotations of the pelvis.

Results from previous lifting studies have shown that the requirement of greater trunk flexion angles have mainly been attributed to increases in pelvic flexion (Potvin *et al.*, 1991; Ursulak and Potvin, 1994)

Chapter 2

REVIEW OF LITERATURE

The Erector Spinae Muscle Group

Forward flexion and extension of the trunk from the flexed position are movements during which the erector spinae muscle group have their most important function (Bogduk and Twomey, 1991). The erector spinae is a large, powerful musculotendinous mass located on the posterior side of the trunk. Superficially, this muscle group can be seen as forming the prominent contours on either side of the posterior aspect of the spine. It consists of three main muscle masses, from medial to lateral: spinalis, longissimus, and iliocostalis. Each of these groups of muscles is capable of several possible actions. No movement of the vertebral column is unique to only one of the three groups of muscles. It is for this reason that these muscle groups can be functionally considered as one. Bilateral activity of the erector spinae muscle group is primarily responsible for extension of the vertebral column. Since origins and insertions of the three muscle groups overlap, entire regions of the vertebral column are moved in a coordinated fashion.

While the spinalis muscle has only cervical and thoracic components, both the longissimus and iliocostalis muscles span the entire presacral vertebral spine. Activity levels of the thoracic and lumbar portions derived by the longissimus and iliocostalis muscles were under investigation in the present study. The proceeding anatomical and functional descriptions of these portions of the erector spinae group have been adopted from Bogduk and Twomey (1991), Oliver and Middleditch (1991), and Warfel (1993).

Thoracic Erector Spinae (TES)

The thoracic erector spinae are composed of thoracic components of both the longissimus and iliocostalis muscles: longissimus thoracis pars thoracis (LT-T) and iliocostalis lumborum pars thoracis (IL-T) respectively (Figure 1). The LT-T originates from the lumbar and sacral spinous processes, and the sacrum between the spinous process of the third sacral vertebra and the posterior superior iliac spine. Muscle fibres pass cranially, with tendons inserting onto the transverse processes of the 1st to 12th thoracic vertebrae and between the tubercle and angle of the 2nd to 12th ribs. The IL-T arises from the superior borders of the angles of the 7th to 12th ribs, and inserts onto the inferior borders of the angles of ribs 1 to 6.

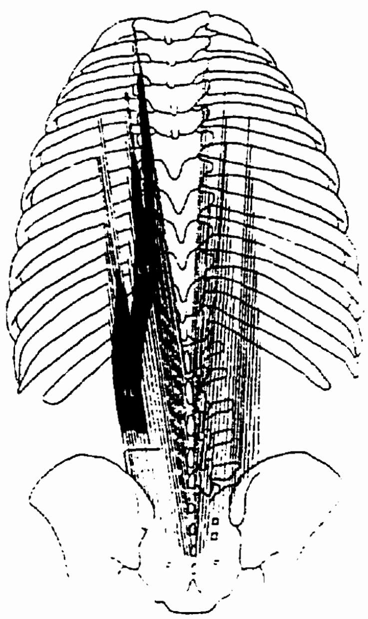

Figure 1. Illustration of the attachments of the Thoracic Erector Spinae musculature (LT-T and IL-T) and their lines of action over the posterior surface of the thoracolumbar spine. From Bogduk and Twomey, 1991.

The TES act principally to extend the thorax over the lumbar spine. They have no direct action on the lumbar spine. Their long tendons of insertion allow these muscles to act around the convexity of the thoracic kyphosis, and anchor the thorax to the ilium and sacrum.

Lumbar Erector Spinae (LES)

The lumbar erector spinae are composed of lumbar components of both the longissimus and iliocostalis muscles: longissimus thoracis pars lumborum (LT-L) and iliocostalis lumborum pars lumborum (IL-L) respectively. The LT-L fibres arise from the accessory and transverse processes of all lumbar vertebrae , and insert onto the ilium, posterior superior iliac spine and sacroiliac ligament. The IL-L fibres originate from the tips of the transverse processes of L_1 to L_4, and from the thoracolumbar fascia. Muscle bellies proceed inferiorly to insert onto the iliac crest and the posterior aspect of the posterior superior iliac spine (Figure 2).

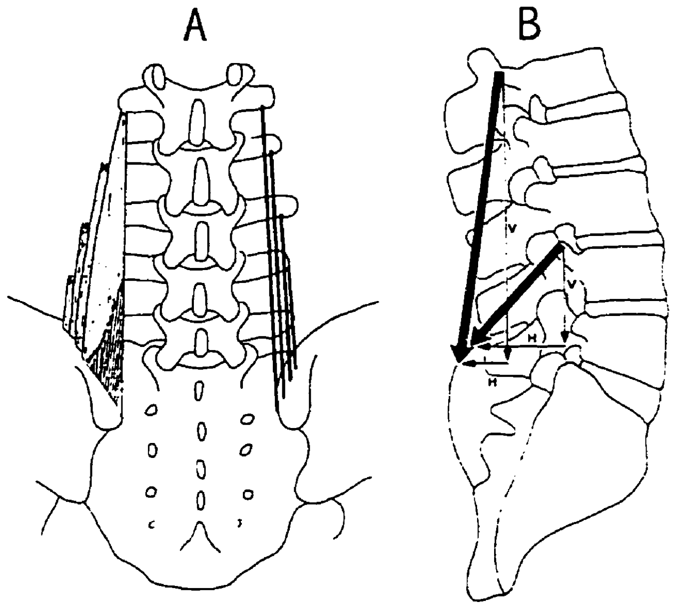

Figure 2. (A) Frontal plane, posterior view of the lumbar spine: An illustration of the attachments and span of the Lumbar Erector Spinae musculature (LES). (B) Sagittal plane view of the lumbar spine: An illustration of the LES lines of action separated into inferior (V), and posterior (H) force vectors. Posterior vectors increase in magnitude at lower lumbar levels. From Bogduk and Twomey, 1991.

Direct attachments of the LES to the lumbar spine, sacrum and ilium allow them to actively manipulate the lumbar lordotic posture. The line of action of the LES is shown in Figure 2. Note that the LES line of action can be resolved into posterior and inferior vectors. Bilateral contraction of the LES will therefore result in two actions of the lumbar spine:

1. The inferior vector (V) will cause posterior sagittal rotation (extension) of the lumbar spine. 2. The posterior vector (H) will exert a posterior translational force on the lumbar vertebrae, resisting against anterior translational forces being imposed on them. The posterior translational force magnitude is greatest at lower lumbar levels where the fascicles of both the LT-L and IL-L assume a greater dorsoventral orientation.

Erector Spinae Aponeurosis (ESA)

The erector spinae aponeurosis is a broad sheet of tendinous fibres that is attached to the ilium, sacrum, and the lumbar and sacral spinous processes. It is formed almost exclusively by the caudal tendons of the longissimus thoracis pars thoracis and the iliocostalis lumborum pars thoracis (TES) (Bogduk, 1980; Macintosh and Bogduk, 1986). The medial half is formed by the longissimus, while the lateral half is formed by the iliocostalis. The lumbar portions of both the longissimus and iliocostalis (LES) have no attachment to the ESA. Thus, with the lumbar portions of these muscles free to move underneath the overlying ESA, this suggests that the LES can act independently from the rest of the erector spinae.

EMG - Force Relationship

The etiology of muscle contractions lies within the human central nervous system. Signals originating from this system are of an electrical form and are termed *action potentials* (Guyton, 1991). Action potentials are defined as rapid changes in membrane resting negative potential to a positive potential, and then back again to the negative potential (Guyton, 1991). It is the conduction of this electrical event from nerve fiber endings to the sarcolemma of a muscle that initiates muscle contraction (Guyton, 1991). A relationship exists between the EMG signal and the development of muscle tension as the EMG recording is a representation of the electrical events at the level of the sarcolemma (Acierno *et al.*, 1995).

The amplitude of the recorded EMG signal is dependant on the spatial and temporal summation of a muscle's motor units within the collection area of the recording electrodes (Fuglevand *et al.*, 1993). These two types of electrical summation are used in the development of muscle tension. The recruitment of motor units within a muscle occurs according to the size principle (Henneman *et al.*, 1965). This principle is based on the observations that for low force contractions, smaller motor units are recruited. As more force is needed, more motor units are recruited in order of size, with the largest ones being the last to be recruited. The recruitment of more motor units creates a spatial summation of the electrical activity within the muscle and leads to higher amplitudes in the EMG signal (Fuglevand *et al.*, 1993). Once a motor unit has been recruited, an increased rate of stimulation to this unit also results in the development of more muscle force. Increasing the rate of stimulation to a motor unit causes an overlap in it's action potentials, causing a temporal summation of the electrical activity. This overlap of electrical activity also leads to increased amplitude in the observed EMG signal (Fuglevand et al., 1993).

EMG - Force relationships have been studied for many muscles under isometric conditions. Linear increases in EMG amplitude with increasing force development have been reported for the forearm muscles (Lind and Petrofsky, 1979), handgrip muscles (Petrofsky *et al*., 1982), and the first dorsal interosseous muscle (Lawrence and De Luca, 1983; Woods and Bigland-Ritchie, 1983). Non-linear relationships have been illustrated for the biceps brachii (Petrofsky *et al*., 1982) and deltoid muscle (Lawrence and De Luca, 1983; Woods and Bigland-Ritchie, 1983). While these studies have provided a basis for the interpretation of EMG signals collected under static conditions, myoelectric analyses under dynamic conditions require further understanding of muscle architecture and how certain biophysical relationships affect their force generating capacity. The force-length relationship illustrates a muscle's force generating capacity at different lengths, with the maximum being obtained at a muscle's resting length (Banus and Zetlin, 1938). The muscle force-velocity relationship indicates the magnitude of force generation to increase as concentric contraction velocities decrease (Hill, 1938). Muscles are capable of generating even greater forces during eccentric contractions. While these relationships represent the force generating capacity of a muscle at a given length or velocity, it should be noted that the actual force developed also depends on it's activation level.

Activity Levels of the Erector Spinae

Spine Forward Flexion

Forward flexion of the spine is primarily facilitated by the force of gravity, but it's rate is controlled by the eccentric contraction of the back extensors (Bogduk and Twomey, 1991; Oliver and Middleditch, 1991). EMG levels of the erector spinae have been shown to increase as the spine flexes forward from the erect position (Allen, 1948; Floyd and Silver, 1951; 1955; Portnoy and Morin, 1956; Carlsoo, 1961; Morris *et al.*, 1962; Okada, 1970; Donish and Basmajian, 1972; Koreska *et al.*, 1977; Ortengren and Andersson, 1977; Andersson *et al.*, 1977). Myoelectric studies of the erector spinae muscle group have shown that this increase is directly proportional to the spine's angle of flexion (Allen, 1948; Andersson *et al.*, 1977 (I, II); Ortengren and Andersson, 1977; Ortengren *et al.*, 1978). This phenomenon can be explained through basic biomechanics. As the angle of spine flexion increases, so does the moment arm, and subsequently the moment of force of the upper body weight about the intervertebral joints (Farfan, 1975). In order to resist the the forward flexion moment of the upper body, there must exist a corresponding extensor moment. Since the change in length of the moment arms of the back extensor muscles to the intervertebral joints are negligible, subsequent increases in their activity levels are required in order to match the increasing forward flexion moment.

Spine Extension from the Flexed Position

Activity levels of the erector spinae muscles during extension of the spine are based on the same biomechanical principles as described for flexion of the spine, noting that extension begins at larger spine flexion angles and ends at erect stance. It has been shown that erector spinae EMG levels are notably greater during extension of the spine as compared to flexion (Allen, 1948; Pauly, 1966; Ortengren and Andersson 1977). This fact can be attributed to a combination of two main factors. The first involves the direction in which the erector spinae muscles are contracting. It has been proven by Hill (1938) that muscles are capable of generating a greater amount of force during an eccentric contraction (muscle lengthening) as opposed to a concentric contraction (muscle shortening). Thus at any given level of force, a muscle will be active at a lower level during an eccentric versus a concentric contraction. Consequently, the erector spinae muscles contract eccentrically during spinal flexion, and concentrically during spinal extension.

The second factor associated with greater levels of erector spinae activity during spinal extension is the force required to overcome that of gravity (Pauly, 1966). While forward flexion of the spine is accomplished in the same direction as the force of gravity, spinal extension from a flexed position is completed in the opposite direction. Thus, greater activity levels are required from the erector spinae muscles during spinal extension in order to oppose the force of gravity acting on the upper body.

During extension of the spine, the combination of a concentric contraction, and having to overcome the force of gravity acting upon the upper body results in greater levels of activation required of the erector spinae muscles as compared to those during spinal flexion.

Control of Pelvic Anteroposterior Rotation

While the erector spinae muscle group is predominantly active during flexion and extension of the spine during lifting and lowering, it also plays an active role in controlling anteroposterior rotation of the pelvis under similar conditions (Oliver and Middleditch, 1991). When acting from attachments with the thoracolumbar spine, increases in activity levels of the erector spinae muscle group will cause anterior rotation of the pelvis on the hip joints (pelvic forward flexion), that will consequently result in an increase in the lordotic posture of the lumbar spine (Figure 3). While the gluteus maximus, hamstrings, and abdominal muscles act to rotate the pelvis posteriorly (pelvic extension), these muscles therefore work antagonistically to the erector spinae in the maintenance of pelvic inclination on the femoral heads. Thus, decreases in activity of the erector spinae will consequently allow the pelvis to extend from a more flexed position. As a result, pelvic extension causes an increase in flexion of the lumbar spine, reducing its natural lordotic posture (Figure 3).

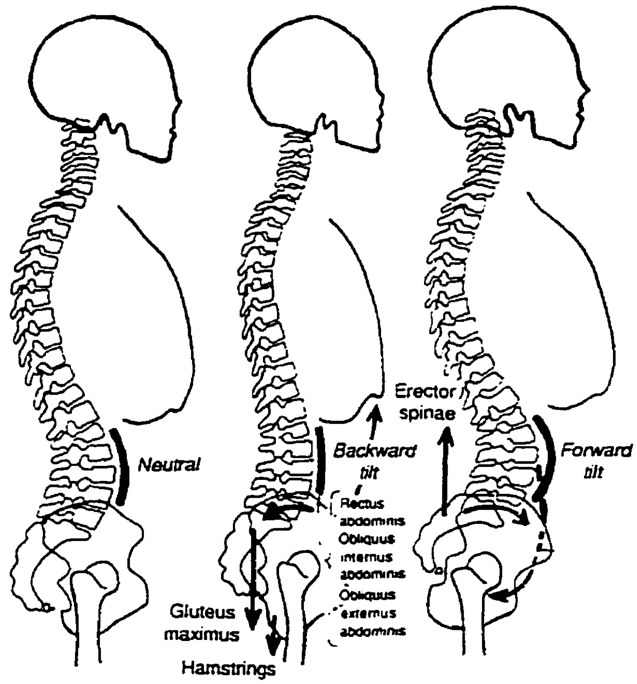

Figure 3. Sagittal plane view of Neutral, Backward tilt (extension), and Forward tilt (flexion) of the pelvis. Backward tilt is caused by activity of the Gluteus Maximus, Hamstring, and Abdominal muscles. Forward tilt is caused by activity of the Erector Spinae musculature. The natural lordotic posture of the lumbar spine in the Neutral position changes oppositely with that of the pelvis. From Oliver and Middleditch, 1991.

Normative Values of Lumbar Spine Flexion

Many studies have been conducted to determine maximum lumbar spine flexion angles in healthy subjects. Mayer and colleagues (1984) used the "two inclinometer technique" on 13 subjects (7 male, 6 female) to determine mean maximum lumbar spine flexion angle of 55 ± 9.2°. Esola et al. (1996) tracked the movement of skin surface diodes on 21 subjects (13 male, 8 female) in the determination of mean maximum lumbar spine flexion of 43 ± 10.3°. Table 1 contains results of studies in the determination of sex specific lumbar spine maximum flexion angles.

Table 1. Summary of results from several studies analyzing maximum lumbar spine flexion angles. *Technique* = equipment used to track lumbar spine motion. *n* = number of subjects. *Mean* = mean maximum lumbar flexion of all subjects. *SD* = standard deviation about the mean. *Range* = range of maximum lumbar flexion across all subjects. (N/A = data not available).

Source	Technique	Sex	n	Mean	SD	Range
Potvin *et al.* (1991)	Watsmart	Male	9	60.2	7.1	N/A
Dolan *et al.* (1994)	3Space Isotrak	Male	23	53.3	7.7	38.5 - 69.4
Dolan *et al.* (1994)	3Space Isotrak	Female	126	56.8	8.9	33.7 - 74.8
Nelson *et al.* (1995)	3Space Isotrak	Female	30	54.7	N/A	36.0 - 74.0

Flexion-Relaxation Phenomenon (FRP)

As previously explained, during progressive spine forward flexion, activity levels of the erector spinae muscle group increase accordingly. Erector spinae muscular activity has been reported to cease at what has been termed as the "critical point" (Floyd and Silver, 1951; Morris *et al.*, 1962; Kippers and Parker, 1984; Kippers and Parker, 1985). Experimental research has shown that it is the LES that experience electrical silence at this critical point (Floyd and Silver, 1955; Portnoy and Morin, 1956; Pauly, 1966; Kippers and Parker, 1984; Kippers and Parker, 1985; Potvin *et al.*, 1991; Wolf *et al.*, 1991; Dolan *et al.*, 1994; McGill and Kippers, 1994; Shirado *et al.*, 1995; Toussaint *et al.*, 1995), while the TES musculature remain active (Potvin *et al.*, 1991; McGill and Kippers, 1994; Toussaint *et al.*, 1995). It has been proposed that the cessation of activity in the LES musculature is due to the substantial deformation in passive tissues on the posterior surface of the lumbar spine, which allows these passive tissues to generate the required extensor moment (Floyd and Silver, 1951; Farfan, 1975; McGill, 1988; Potvin *et al.*, 1991; McGill and Kippers, 1994). While Allen (1948) first discovered this phenomenon, it was Floyd and Silver (1951) that coined the term "flexion relaxation".

Onset of the FRP

Unloaded Trunk Movement

Some scientists studying the onset of FRP during unloaded trunk forward flexion have claimed that the termination of LES activity, when observed, occurs at maximum spine flexion (Allen, 1948; Floyd and Silver, 1951; Floyd and Silver, 1955). Others have observed the occurrence of this phenomenon at maximum trunk flexion (Morris *et al.*, 1962; Pauly, 1966). Some studies have also detected the onset of the FRP prior to full trunk flexion (Portnoy and Morin, 1956; Okada, 1970; Wolf et al., 1979; Wolf *et al.*, 1991; Shirado et al., 1995). As trunk flexion is a multisegmental motion, accomplished by flexion of both the spine and pelvis, and due to the etiology of the FRP as aforementioned, the occurrence of this phenomenon should be cited with reference to flexion angles of the spine and not the trunk. This claim is supported by Kippers and Parker (1984) in which they obtained a high correlation ($r = 0.95$) between spine flexion angle and (a) the silencing of the LES during unloaded trunk flexion and (b) the recommencement of LES activity during unloaded trunk extension. Conversely, the correlation between trunk flexion angle and these two events was observed to be low ($r = 0.46$) (Kippers and Parker, 1984). In addition to its high correlation, spine flexion angle had considerably lower variability than trunk flexion angle at the onset of the two FRP events (6.4° vs. 12.8°) (Kippers and Parker, 1984).

Loaded Trunk Movement

The onset of the FRP has been observed to occur at greater lumbar spine flexion angles during loaded trunk movement than those observed during unloaded conditions. Kippers and Parker (1984) performed a study in which the goal was to determine the spine flexion angles (expressed in degrees and as percent of maximum forward flexion) at which the cessation of LES activation occurred during the lower (SP1), and also upon the reactivation of LES during the lift (SP2). Eleven subjects were required to perform a stoop lower and lift of 10.1 kg. While holding the load, each subject was instructed to maximally flex the trunk from the erect position, momentarily maintain full flexion, and then extend back to the erect position. Spine flexion angle at SP1 and SP2 were 52.1° (95.6% max.) and 52.6° (96.5% max.) respectively.

Sagittal Plane Trunk Motion: Contributions from the Lumbar Spine and Pelvis

It is generally accepted that trunk flexion (θ_T) is accomplished via a combination of spine flexion (θ_S) and that of the pelvis about the hip joints (θ_P) (Davis *et al.*, 1965; Farfan, 1975; Gracovetsky *et al.*, 1977; Kippers and Parker,1984; Mayer *et al.*, 1984; Potvin *et al.*, 1991; Paquet *et al.*, 1994; Nelson *et al.*, 1995; Esola *et al.*, 1996), such that $\theta_T \approx (\theta_S+\theta_P)$. Studies have shown this same relationship to exist during trunk extension (Farfan, 1975; Kippers and Parker, 1984; Anderson *et al.*, 1985; McGill and Norman,1986; Potvin *et al.*, 1991; Paquet *et al.*, 1994; Nelson *et al.*, 1995; McClure *et al.*, 1997). It is important to note in studies analyzing sagittal movement of the thoracolumbar spine that although it is considered a single structure, collected data reflect mainly lumbar motion as mobility in the sagittal plane generally decreases from the lower lumbar to the upper thoracic regions (Tanz, 1953; Allbrook, 1957; White and Panjabi, 1975; White and Panjabi, 1978). While an extensive number of studies have been performed on the analysis of the pattern of angular displacement between the lumbar spine and pelvis throughout unloaded sagittal plane trunk movements, very few have been conducted under conditions in which subjects handled a load. There currently exist two theories depicting the pattern of lumbar spine and pelvic movement during sagittal plane trunk motion: (a) they move in a synchronic pattern and (b) they move in a sequential pattern. Those who support the sequential strategy claim that from erect stance, early sagittal trunk flexion is achieved more exclusively through the lumbar spine, and that later stages of trunk flexion are provided almost solely by pelvic rotation (Farfan, 1975; Gracovetsky *et al.*, 1977; Kippers and Parker, 1984; Mayer *et al.*, 1984; Paquet *et al.*, 1984; Esola *et al.*, 1996). It has also been proposed that this pattern of lumbar spine and pelvic movement is reversed when extending from a forward flexed posture (Farfan, 1975; Gracovetsky *et al.*, 1977; McClure *et al.*, 1997). Synchronicity of lumbar spine and pelvic sagittal motion implies that they are moving simultaneously (Potvin *et al.*, 1991; Burgess-Limerick *et al.*, 1992; Nelson *et al.*, 1995), with one of the two possibly flexing/extending at a greater velocity during particular stages of trunk movement.

Unloaded Conditions

Several techniques have been employed in the determination of the pattern of movement between the lumbar spine and pelvis during unloaded sagittal plane motion. Kippers and Parker (1984) simply plotted sagittal plane flexion of the lumbar spine, pelvis, and trunk during unloaded forward flexion motions. Paquet *et al.* (1994) plotted pelvis versus spine normalized angular displacements in describing their pattern of movement. Certain kinematic studies calculated lumbar spine - to - pelvic flexion ratios throughout designated phases of trunk forward flexion (Mayer *et al.*,1984; Esola *et al.*, 1996), and trunk extension from a flexed position (McClure *et al.*, 1997). Despite the various techniques used to study the pattern of lumbar spine and pelvis motion, all concluded that their motion is accomplished sequentially.

It must be noted that tasks involving lifting loads account for the majority of occupationally related risks of low back disorder (Bigos *et al.*, 1986; Spengler *et al.*, 1986; Snook, 1989; Videman *et al.*, 1990). Hence, the study of the pattern of movement between the lumbar spine and pelvis under conditions of lifting and lowering loads would be more relevant to occupationally related injury risk.

Loaded Conditions

(a) Kinematic Studies

Very limited data has been collected to date on the description of the lumbar spine - pelvic sagittal plane pattern of movement while manually handling loads. A kinematic and electromyographic sagittal plane study performed by Potvin and colleagues (1991) analyzed pelvic and lumbar spine movements when subjects were lifting loads. Visual analysis of absolute spine flexion versus pelvic flexion plots led them to conclude that these two structures move in a synchronous pattern while lifting a load (Potvin *et al.*, 1991). In support of the results of Potvin *et al.* (1991), Burgess-Limerick and associates (1992) attempted to quantify the synchronous movement of the lumbar spine and pelvis during lifting via (a) plotting absolute flexion angles of the lumbar spine versus the pelvis and (b) calculating relative phase angles between these two segments. The instantaneous phase of each structure at each point in time was calculated as the arc tangent of the ratio of normalized velocity to normalized displacement, and the relative phase angles were calculated as the difference in these values between the two segments (Burgess-Limerick *et al.*, 1992). This approach appears to have merit, however the results of their pilot study appeared to be unreliable due to noisy data and too few subjects (n=3). Nelson *et al.*(1995) completed a study in which a 3Space Polhemus Tracker System was used to measure differential lumbar spine and pelvic motion during stoop lifting (loaded extension) and stoop lowering (loaded flexion) conditions. Inspection of slopes of normalized lumbar spine and pelvic flexion angles plotted against normalized gross spinal flexion angles led these scientists to believe that a synchronous pattern of lumbar spine and pelvic flexion exists during loaded flexion. However, they observed that trunk extension during lifting is completed with a more sequential strategy; with pelvic extension dominating at greater trunk angles, and lumbar extension providing most of trunk movement when approaching more erect posture (Nelson *et al.*, 1995).

(b) Biomechanical Modelling

Biomechanical models have also calculated sagittal plane angular displacements of the lumbar spine and pelvis during lifting through mathematical algorithms (Farfan, 1975; Farfan and Lamy, 1977; Gracovetsky *et al.*, 1977; Anderson *et al.*, 1985). Over the past two decades, biomechanists have attempted to develop mathematical models of the lumbosacral joint during lifting activities in order to predict low back tissue stress. Outcomes of these studies have also resulted in predictions of whether lumbar spine and pelvic movement are synchronous or sequential during lifting tasks. Farfan (1975) reported that during trunk flexion, the intial 60° is accomplished almost exclusively by flexion of the lumbar spine. It was also proposed that trunk extension during lifting is accomplished sequentially, with pelvic extension solely initiating movement and lumbar spinal extension taking over at smaller trunk flexion angles (Farfan, 1975). This was illustrated by plotting lumbar flexion angles at which muscle pull or ligament tension counteract the forward flexion moment imposed by the upper body and specific additional loads enough that lumbar spinal extension could occur. The rationale behind this theory was that the gluteal and hamstring muscles together have a far greater cross sectional area and a longer moment arm than the erector spinae and as a result are capable of generating substantially larger extensor moments (Farfan, 1975). Thus, during initial stages of a lift, the gluteal and hamstring muscles contract to extend the pelvis, reducing the moment of the load about the lumbosacral joint to the point that the erector spinae can act to extend the spine (Farfan, 1975). The Lamy-Farfan model of the lumbar spine also calculated the first 40° of trunk flexion to be accomplished solely by the lumbar spine, while the remaining degrees of trunk flexion were accomplished by the pelvis (Farfan and Lamy, 1977). The posterior spinal ligamentous system would take over the required extensor moment at lumbar spinal flexion angles greater than 40° (Farfan and Lamy, 1977). Gracovetsky *et al.* (1977) incorporated more parameters into the Lamy-Farfan model, and explained that due to their longer moment arms, the posterior spinal ligamentous system was more mechanically efficient at generating extensor moments than the muscular system. While handling a load, the required extensor moment for lumbar spine flexion angles between 0° and 40° could be maintained by the muscular system (Gracovetsky et al., 1977). At flexion angles greater than 40°, the extensor requirements exceeded the capability of the muscular system. It was stipulated that pelvic movement was therefore required in order to change lumbar spine geometry, which would bring the ligamentous system into action (Gracovetsky *et al.*, 1977). This biomechanical premise was the basis behind these scientists' conjecture of sequential movement between the lumbar spine and pelvis. Conclusions based on results of the previously described studies are limited in that no analyses of pelvic flexion/extension were made. Thus, no results are available to prove that pelvic extension may be occurring simultaneously with the lumbar spine during lifting.

Anderson *et al.* (1985) used knee angle and trunk flexion angle during trials for the development of regression equations in determining sacral degree of rotation and percent of maximum L_5/S_1 relative rotation. Through graphic representation of these equations, it was determined that the first 30° of trunk flexion is primarily accomplished through flexion of the lumbar spine, but shifts predominantly to pelvic flexion at trunk angles beyond this value (Anderson *et al.*, 1985). Reliability of these results are questionable as standard error for the sacral and L_5/S_1 rotation equations were 7.4% and 16.4% respectively (Anderson *et al.*, 1985). This renders standard deviations to be very large.

Manual Materials Handling: Peak Lumbar Spine and Pelvic Flexion Angles Associated with Sagittal Plane Trunk Motion

(a) Squat vs. Stoop

Two main types of lifting have been compared to date: squat and stoop. Squat lifts are accomplished by flexing primarily at the knees, while stoop lifts are performed mainly by flexing the trunk and maintaining a more extended knee angle. Several studies have found significant differences in peak lumbar spine and peak pelvic flexion angles between lifting techniques (Potvin *et al.*, 1991; Ursulak and Potvin, 1994). While there is evidence of an increase in peak lumbar spine flexion from squat to stoop lifting, increments have been reported as being substantially less than that of pelvic flexion (Potvin *et al.*, 1991; Ursulak and Potvin, 1994). Potvin *et al.* (1991) observed an average peak trunk flexion angle of 64° for squat lifts, and 112° for stoop lifts. While there was a dominant contribution of the lumbar spine to the first 40° of trunk flexion during the squat lifts, only 11° of the 48° increase in trunk flexion during the stoop lifts was due to further lumbar spinal flexion (Potvin *et al.*, 1991). Thus, the increase in peak trunk flexion angle during stoop lifts was attributed mainly to a 37° increase in peak pelvic flexion (Potvin *et al.*, 1991). Ursulak and Potvin (1994) obtained strikingly similar results to Potvin *et al.* (1991) when comparing peak trunk, lumbar spine, and pelvic flexion angles during squat and stoop lifts. Average peak trunk flexion for the squat and stoop lifts were 48.1° and 102.2° respectively (Ursulak and Potvin, 1994). Peak lumbar spine and pelvic flexion angles for the squat lifts were 40.2° and 19.2° respectively (Ursulak and Potvin, 1994). In contributing to the 54.1° increase in peak trunk flexion angle from squat to stoop lift, peak lumbar spine flexion increased by only 13.1°, while peak pelvic flexion increased 31.9° (Ursulak and Potvin, 1994). These two studies indicate that trunk flexion during squat lifts is accomplished mainly through lumbar spine motion, and although peak lumbar flexion increases during stoop lifts, the increase in peak trunk flexion is accomplished predominantly via increases in peak pelvic flexion.

(b) Lift vs. Lower

Very limited data has been collected on lift/lower differences in peak lumbar spine flexion. DeLooze *et al.* (1993) analyzed for lift/lower differences in peak L_5/S_1 flexion angles, but with a limited subject pool (n = 8). Large amounts of data have been collected in psychophysical studies in order to determine maximum acceptable weights of lift and lower for both sexs (Snook and Ciriello, 1991). Development of tables for determining these values illustrate higher weight limits for lowering than for lifting

(Snook and Ciriello, 1991). Kinematic investigations are needed for comparisons of lifting and lowering tasks in order to determine whether differences exist in injury risk.

(c) Sex Differences for any conditions

Very limited work has been done in the domain of sex differences in peak lumbar spine and peak pelvic flexion angles during any style of lifting and lowering. Dolan *et al.* (1994) performed a study in which sex comparisons were made for peak lumbar spine flexion and passive tissue contributions to extensor moment during three variations of lifting. No sex differences were observed for peak lumbar spine flexion within any lifting condition. At a given percent of maximum lumbar spine flexion, while it was calculated that males had greater absolute passive tissue contributions to back extensor moment, no sex differences existed in these values relative to the peak extensor moment being generated (Dolan *et al.*, 1994).

Chapter 3

METHODOLOGY

Subjects

Eleven male and eleven female subjects volunteered to participate in this study. Anthropometric data are presented in Table 1. Only subjects with no history of low back injury or recurring pain were used in this study. Informed consent was obtained from all participants. A copy of the consent form is included in Appendix A.

Table 2. Anthropometric data (age, height, and weight) presented as means ± standard deviations for both males and females, and for all subjects combined.

Sex	Age (years)	Height (cm)	Weight (kg)
Male (n=11)	27.4 ± 3.0	176.4 ± 5.7	78.2 ± 10.3
Female (n=11)	25.8 ± 3.4	162.2 ± 3.7	54.6 ± 5.1
Overall (n=22)	26.6 ± 3.3	169.3 ± 8.6	66.4 ± 14.5

Collection Session

The data collection session was approximately 2 hours in duration, which consisted of subject and equipment set-up, acquisition of muscular maximum voluntary contractions (MVC's), and the completion of experimental load handling trials.

Collection Equipment

A portable electromyography (EMG) system (biovision) was used for the collection of muscular activity levels. Pre-amplifier gain was set at 1000, input impedance was 10^{12} Ω, and CMRR was 120 dB. EMG data were collected and processed with LabVIEW 3.1 (National Instruments, Austin, Texas, USA) using a MacIntosh IICi computer and a 12 bit multifunction I/O board (NB-MIO-16X, National Instruments). Both MVC and experimental trial EMG data were subjected to the same data processing. Signal collection specifications were as follows: sampling rate of 1020 Hz, bandwidth of 15 - 450 Hz, full wave rectified, low-pass filtered using a 1st order Butterworth filter with a cut-off frequency of 2.5 Hz. Data were then converted to 30 Hz before saving to file.

Two dimensional sagittal plane videography was used to record kinematics for all trials. Videography was performed using a Panasonic System Camera (model #: WV-D5000) at 60 Hz, and was recorded onto Super VHS vidoetape using a Panasonic video cassette recorder (model #: AG-2400).

Synchronization of videography and EMG was accomplished by the use of a trigger that simultaneously lit up an LED in the camera field of view, and initialized EMG collection. The LED was instantly triggered at the start of every trial.

Subject Preparation

Prior to data collection, each subject had reflective kinematic markers and two pairs of surface electrodes (Ag-AgCl bipolar electrodes: Medi-Trace disposable ECG electrodes, Graphic Controls, Gananoque, Ontario, Canada) affixed to their skin. Anatomical location of the reflective markers were defined as follows (Figure 4): Ankle: lateral malleolus (of fibula), Knee: lateral condyle of femur, Hip: greater trochanter of femur, L_4/L_5: iliac crest directly superior to hip marker, Shoulder: lateral edge of acromion process of scapula, Elbow: lateral condyle of humerus, Knuckle: metacarpophalangeal joint of third digit, Fins: 1. C_7/T_1 - 2 marker fin affixed to posterior spinous process of C_7/T_1, 2. Lumbar - 3 marker triangular fin affixed to posterior spinous process of L_1, 3. Pelvic - 3 marker triangular fin affixed to the posterior surface of the sacrum.

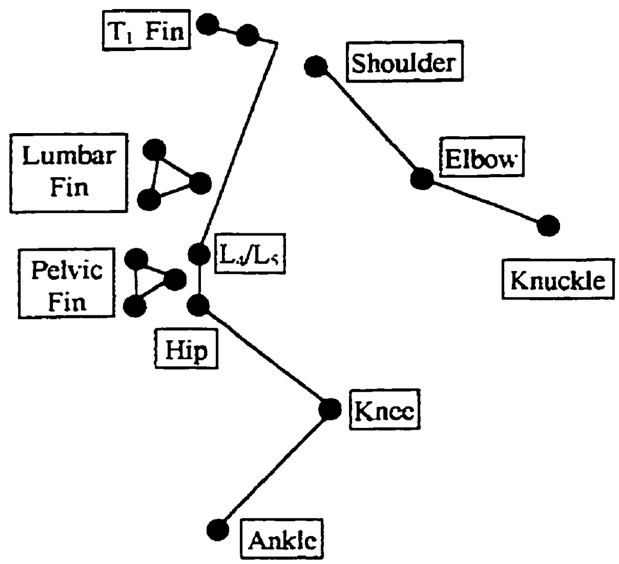

Figure 4. Stick figure illustration of anatomical locations of reflective markers on the surface of the body.

Myoelectric signals were collected from two channels: the thoracic (TES) and lumbar (LES) erector spinae muscles. Unilateral sites were chosen assuming muscle symmetry during the sagittal plane tasks. Specific muscle sites were as follows: TES electrodes were placed 3 cm lateral to the posterior spinous process of the ninth thoracic vertebrae;

LES electrodes were placed 3 cm lateral to the posterior spinous process of the fourth lumbar vertebrae. Electrode pre-amplifiers were taped to the skin of each participant so as to reduce the noise content in the myoelectric signal that may have been created by pre-amplifier movement. A ground electrode was also fixed to the skin over the right anterior ribcage.

Maximum Voluntary Contractions

Acquisition of MVC's from both muscle groups were obtained so that the recorded experimental trial EMG signals could be normalized to maximal activity. MVC's were recorded from subjects via a torso harness and pelvic support apparatus (Figure 5). Subjects put the harness around the torso, with the pelvis supported anteriorly by a concave shell and padding. They were then instructed to gradually extend the back until maximum force generation was achieved by both sets of muscles. During this task, subjects were further instructed to attempt to retract the scapulae around the posterior surface of the thorax in order to get a maximum effort from the TES musculature. Ramping of muscle force generation was employed so as to avoid occurrence of muscle injury. This was accomplished by instructing subjects to perform the MVC trials slowly. Each subject performed three MVC trials and the largest myoelectric output of each muscle was selected as it's maximum. Normalization to percent MVC was accomplished by dividing trial muscle activity levels by their maximum levels, and multiplying by 100%.

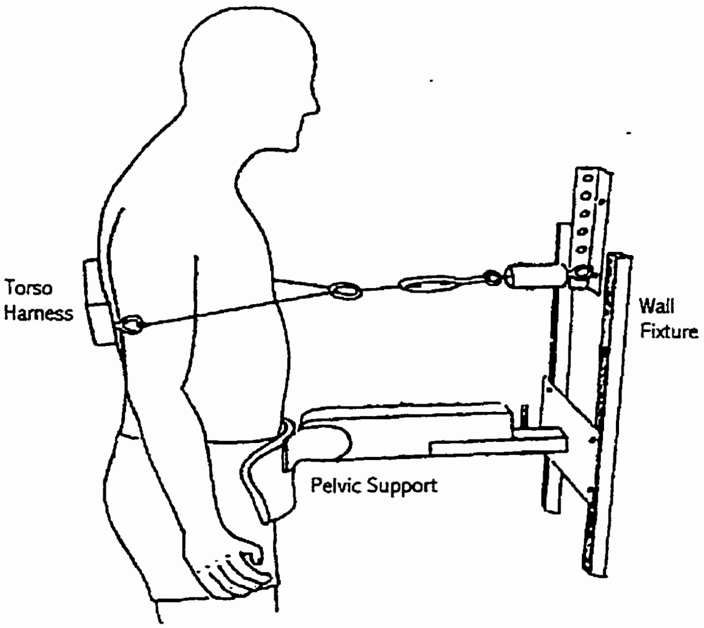

Figure 5. Illustration of the MVC acquisition apparatus, consisting of the subject's torso supported within a harness, and the pelvis supported anteriorly by a concave shell and padding.

Lifting and Lowering Protocol

Subjects began each data collection session by performing three tasks in the camera field of view. With all reflective markers and electrodes on the designated sites, participants were required to stand erect in order to collect standing bias values for the ribcage, pelvis, and knee angles. Subjects were then required to lock their knees in the fully extended position and flex their upper body forward in order to obtain a maximum lumbar spine flexion angle. Finally, each subject was instructed to hold a meter stick vertically in front of themselves for scaling purposes. The meter stick had reflective tape applied to both ends.

Each subject performed 10 lifts and 10 lowers under both a "freestyle" and "constrained" condition. The difference between the two conditions was that an obstacle in the form of a wall was placed between the subject and the load for the "constrained" condition. The wall was designed to simulate an industrial bin. In order to simulate an industrial bin with walls that exceed knee height, wall heights were calculated as 120% of average male and female knee heights. Knee height is documented as being 28.5% of total body height (Chaffin & Anderson, 1991). Average Canadian male and female heights are 174.8 cm and 160.8 cm respectively (Webb Associates, 1978). Thus, wall heights were 60 cm for males, and 55 cm for females in the "constrained" condition. The center of the load was located 20 cm in front of the wall edge nearest the subject, for the start of the first trial for each condition.

Each trial began with the subject in the erect position, holding no load. They flexed forward to pick up the load and returned to the erect position. From this position the subjects flexed forward once again to place the load back to the floor. The lift stage corresponded to the time from which the subject touched the load, until they extended back up to the erect position. The lower began from the erect position while holding the load, until it was placed back onto the floor. Subjects were instructed to perform each trial in a manner most comfortable for them. The load was not moved by the experimenter in between trials in order that the load location on the floor would be the same for the lift and lower.

Load Magnitude

Both male and female load magnitudes were based on psychophysical limits developed by Snook & Ciriello (1991). Materials handling criterion: box width was selected as 49 cm, vertical distance through which the load travelled was between the floor and knuckle height (76 cm), frequency of 4 lifts per minute was selected. These criteria yielded a male maximum acceptable weight of lift and lower of 14 and 16 kilograms respectively. Female load magnitudes were selected as 9 and 8 kilograms respectively. Averages were taken of the lift and lower magnitudes for both sexs to yield a final load magnitude of 15 and 8.5 kilograms for males and females respectively. Lead shot was weighed out to the determined load magnitudes and poured into a bag. This was done to avoid shifting around of lead shot in the lifting box. The bag of lead shot was then placed into the lifting box.

Data Treatment

Kinematics

(a) Digitizing

Two dimensional digitizing of the videotaped data was conducted in Peak5 software, version 5.2.1 (Peak Performance Technologies Inc., Englewood, Colorado, USA). This software also used the meter stick in the camera field of view in order to develop a scale factor. Meter stick end points were manually digitized and identified as being one vertical meter apart. Digitizing of the end points was averaged over three trials. Scale factors were used in future data processing in order to digitally convert raw units to metric units. The coordinates for all markers were digitized from each second frame, resulting in a rate of 30 Hz.

The start of each digitized trial was selected as the frame in which the LED first lit up. Frame digitizing ceased when the experimenter could visually see the subject letting go of the load at the end of the lower. At this point, all raw coordinates were stored onto the hard drive of a PC compatible computer. Digitizing using the coordinates shown in Figure 1 was performed for all trials of both freestyle and constrained conditions. These coordinates were also used for the digitizing of the erect stance, and maximum lumbar spine flexion trials.

(b) Angle Definitions

Angle calculations were performed on the digitized coordinates for each frame using Microsoft QuickBASIC 4.50 software. Note that due to the nature of these calculations, anatomical flexion resulted in an increase, while anatomical extension resulted in a decrease for each of the angles.

1. Trunk Angle was defined as the trigonometric angle subtended between the L_4/L_5 marker, and the calculated T_1 coordinates within the neck. Standing bias trunk angle was set as -90°. With reference to Figure 6, coordinates for the location of the T_1 vertebral body (x_3, y_3) were determined by equations that extrapolated the vector created between (x_1, y_1) and (x_2, y_2) into the neck.

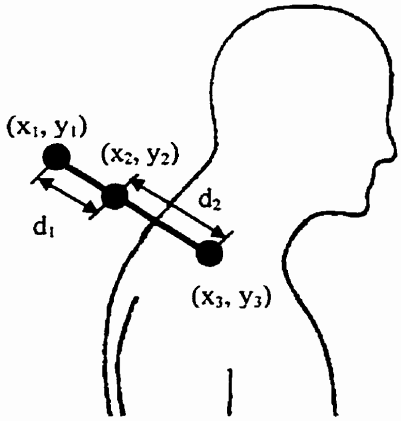

Figure 6. Description of how T_1 location (x_3, y_3) was determined via extrapolation of the vector created between (x_1, y_1) and (x_2, y_2) on the external fin. Note that d_1 is the distance between the fin markers (5.5 cm), and d_2 is the distance from the fin marker 2 to the estimated location of the T_1 vertebral body. These distances were both measured constants.

Thus, calculations of (x_3, y_3) are:

$$x_3 = (8.5/5.5*(x_2 - x_1)) + x_2$$

$$y_3 = (8.5/5.5*(y_2 - y_1)) + y_2$$

It is important to note that d_2 (8.5 cm) was measured out on one subject, and assumed to be constant for all subjects. Differences in d_2 between subjects due to anatomical differences (i.e. levels of subcutaneous fat deep to the anchoring of the C_7/T_1 fin) were assumed to be negligible.

2. Knee Angle was defined as the thigh angle minus the shank angle. Thigh angle was defined as the trigonometric angle subtended between the hip and knee marker. Shank angle was defined as that subtended between the ankle and knee marker. Standing bias knee angle was 0°. Knee angle increased from 0° upon flexion.

3. Ribcage Angle corresponded to the rotation of the lumbar fin.

4. Pelvic Angle corresponded to the rotation of the sacral fin.

5. Lumbar Spine Angle was defined as Ribcage Angle minus Pelvic Angle.

***Descriptions of fin rotations are in Appendix B.**

(c) Treatment

Kinematic data were analyzed with Microsoft QuickBASIC 4.50 software. Due to the time demands of data processing, only the final three trials of each experimental condition were analyzed for each subject. Raw coordinate data were dual pass filtered using a 4^{th} order Butterworth filter with an effective low cut-off frequency of 5 Hz. The first and last 10 frames of each trial were mirrored onto themselves to allow for frames to "warm up" the filter during both passes.

Trunk, knee, pelvis, and lumbar spine angular displacement time-histories were calculated from the filtered raw data, with standing bias angles subtracted. From these angular time-histories, peak flexion angles were averaged over the final three trials per condition. Lumbar spine peak flexion angles were expressed as a percent of maximum ($\theta_{\%LUM}$). This was accomplished by dividing each trial peak lumbar spine angle by the subject's maximum angle of lumbar spine flexion, and then multiplying by 100%.

Instantaneous angular velocities were calculated for the lumbar spine and pelvis throughout each trial via central difference, using the two interval method, where $\omega_2 = (\theta_3 - \theta_1)/2\Delta t$, with ω_2 = angular velocity at frame #2, θ_3 = angular displacement at frame #3, θ_1 = angular displacement at frame #1, and Δt = time difference between two successive frames. Angular velocities were then dual pass filtered using a 4^{th} order Butterworth filter with an effective low cut-off frequency of 2.5 Hz. Both lumbar spine and pelvic angular displacements and velocities were used for the calculation of phase angles for their corresponding anatomical structure. The instantaneous phase of each structure was calculated as the arc tangent of the ratio of normalized velocity to normalized displacement. Lumbar spine and pelvis calculations were performed in the exact same fashion so a description of calculations is provided for the lumbar spine only. Within each trial, lumbar spine angular displacements greater than zero were normalized by dividing them by the maximum positive angular displacement, while those less than zero were normalized by dividing them by the absolute value of the maximum negative angular displacement. Thus, the maximum and minimum normalized angular displacement of each trial was +1 and -1 respectively. Angular velocities were positive during the lower and were therefore normalized by dividing them by the maximum positive angular velocity. This yielded a maximum angular velocity of +1 for the lower. Angular velocities for the lift were negative and thus were all divided by the absolute value of the maximum negative angular velocity. This yielded a maximum angular velocity of -1 for the lift.

A relative phase angle was then calculated as the pelvis phase angle subtracted from that of the lumbar spine throughout the duration of each trial. Refer to Appendix E for graphic representation of the steps taken in the calculation of instantaneous relative phase angles for a typical trial of a constrained lower. These calculations were originally derived by Burgess-Limerick *et al.* (1992).

Two types of hysteresis plots were also collected from each trial: pelvis flexion angle versus lumbar spine flexion angle (P-S plot), and horizontal distance of load from the ankle (H (cm)) versus vertical distance of the load from the floor (V (cm)) (H-V plot). H was calculated as the difference between the X coordinates of the knuckle marker and ankle marker, while V was calculated as the difference between the Y coordinates of the same markers. Definitions for the horizontal and vertical calculations follow those used for the NIOSH lifting equation (NIOSH, 1991).

Windowing

Trial start and end points were determined by calculating knuckle marker vertical velocity of each subject throughout each trial. This was done in Microsoft QuickBASIC 4.50 software. Starting points were selected as the frame at which the knuckle vertical velocity began to abrubtly shift away from 0 raw units/second, or from a constant velocity that was close to 0 raw units/second (Figure 7). Conversely, end points were selected as the frame at which knuckle vertical velocity returned back to a constant level (Figure 7). The determined start and end points of both the lift and lower of each trial were used for windowing both kinematic and EMG data.

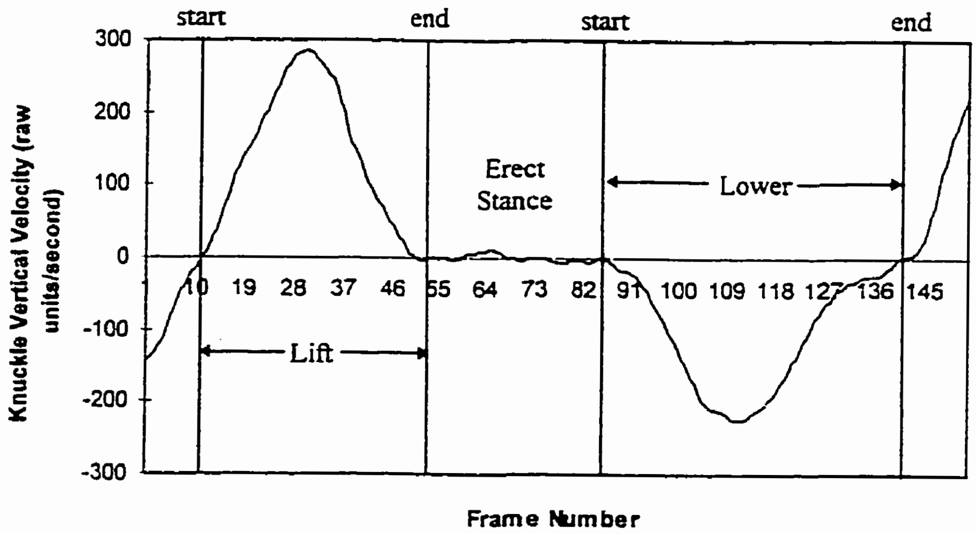

Figure 7. Illustration of knuckle vertical celocity throughout the duration of one trial. Visual inspection of these plots allowed for the determination of frame numbers at which the lift started and ended, therefore designating the frames in between these two as the lift. The same was done for the lower.

EMG

For each subject, signals were windowed according to the lift and lower start and end points. EMG windowing was accomplished using Microsoft QuickBASIC 4.50 software. Within each subject, EMG signal averaging was conducted according to *lifting condition* (freestyle/constrained), and the *direction* of the load (lift/lower), resulting in four combinations per channel: freestyle lift, freestyle lower, constrained lift, and constrained lower. Each combination contained ten trials per subject. Thus for each combination within a subject, averages were taken of the ten peak and mean EMG values.

Statistical Analysis

Maximum lumbar spine flexion angles were compared between males and females using a t-test in StatVIEW II (Abacus Concepts Inc., Berkeley, California, USA). All other statistical analyses were performed in SuperANOVA version 1.11 (Abacus Concepts Inc., Berkeley, California, USA). Post-hoc t-tests were performed in StatVIEW II. A significance of 0.05 was chosen for all tests.

Analysis Model

A repeated measures 2 × 2 × 2 ANOVA was employed for all peak flexion angles (trunk, knee, pelvis, $\theta_{\%LUM}$), mean and peak EMG data (thoracic and lumbar channels). The repeated measures model consisted of two within variables (*lifting condition* and *direction*) and one between variable (*sex*). *Lifting condition* had two levels (freestyle and constrained), and *direction* had two levels (lift and lower). Post-hoc contrasts by means comparisons were used to test for significant differences between interactions of within variables. Post-hoc t-tests were performed on interaction means that included both within and between variables.

Kinematics

Qualitative analyses were performed for the angular displacement time-histories, relative phase angles, pelvis flexion angle versus lumbar spine flexion angle plots, and horizontal distance of load from the ankle versus vertical distance of the load from the floor plots.

These were also averaged over final 3 trials of all subjects for each of the four *lifting condition* × *direction* combinations.

Chapter 4

RESULTS

No data were missing from the collection or analyses and thus results are reported for all 11 males and 11 females, to yield an overall study sample size of n = 22. As kinematic analyses involved averaging over the final three trials of each subject, the total number of trials included were 264. EMG averaging was done over all ten trials per subject, yielding a total of 880 trials for analysis. Unless stated otherwise, data presented in this section are cited as: mean ± 1 standard error. Standard error has been calculated using the **n** indicated within the corresponding figure.

Maximum lumbar spine flexion angles are reported in Table 3. Statistical analyses showed no significant difference between maximum lumbar spine flexion angles for males and females.

Table 3. Statistics for maximum lumbar spine flexion angles pooled within *sex* (n = 11 for each sex), and also across all subjects (n = 22).

Sex	Mean	Standard Deviation	Range
Male	53.2	9.5	40.2 - 73.0
Female	55.2	7.3	47.50- 68.9
Overall	54.2	8.4	40.2 - 73.0

Main Effects

Sex Effect

Statistical analyses illustrated no *sex* effect for all kinematic and EMG variables ($p > 0.05$).

Lifting Condition Effect

Kinematic Variables

Mean peak pelvis, trunk, knee, and percent of maximum lumbar spine ($\theta_{\%LUM}$) flexion angles are illustrated for freestyle and constrained conditions in Figure 8. Both peak pelvis and trunk flexion angles were significantly smaller for the freestyle than the constrained condition. Peak knee angle showed an opposite trend in that it was greater for the freestyle than the constrained condition. No *lifting condition* effect existed for $\theta_{\%LUM}$.

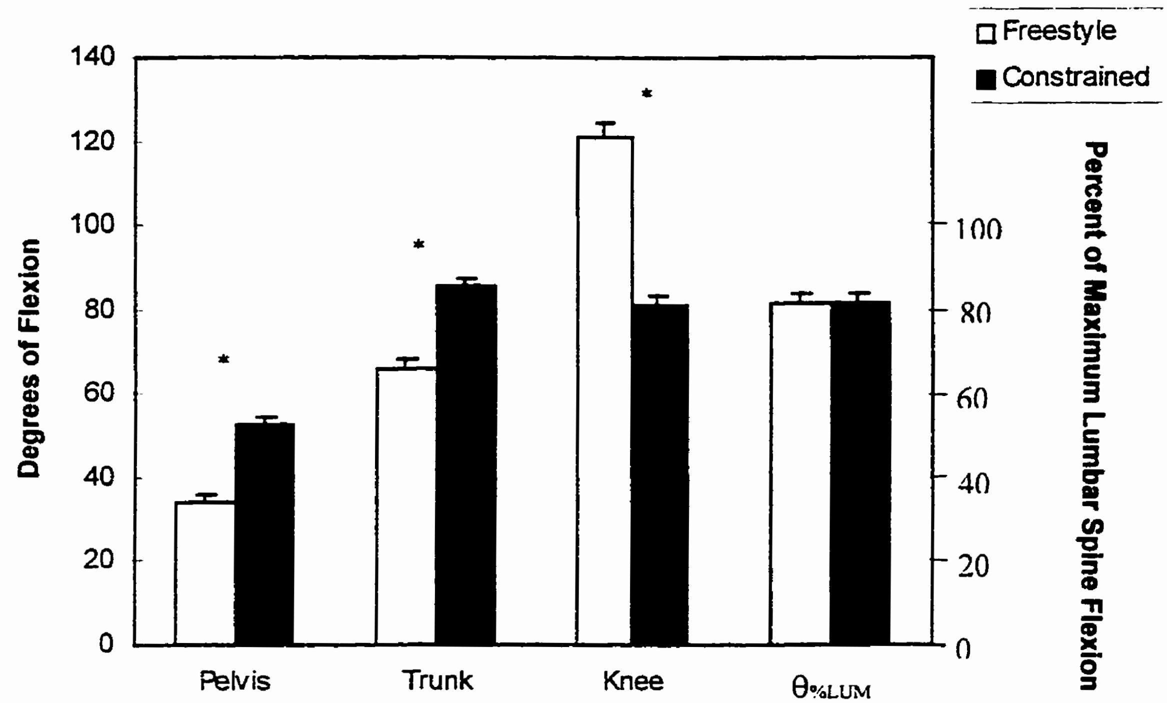

Figure 8. Mean peak Pelvis, Trunk, and Knee flexion angles, and $\theta_{\%LUM}$ for freestyle and constrained conditions. Data are pooled across *sex* and *direction* (n = 44). (* = significant *lifting condition* effect at $p < 0.001$).

EMG Variables

Mean and peak activation levels of both TES and LES muscle groups are shown for both conditions in Figure 9. Freestyle activation levels were significantly lower than those for the constrained condition for all EMG variables.

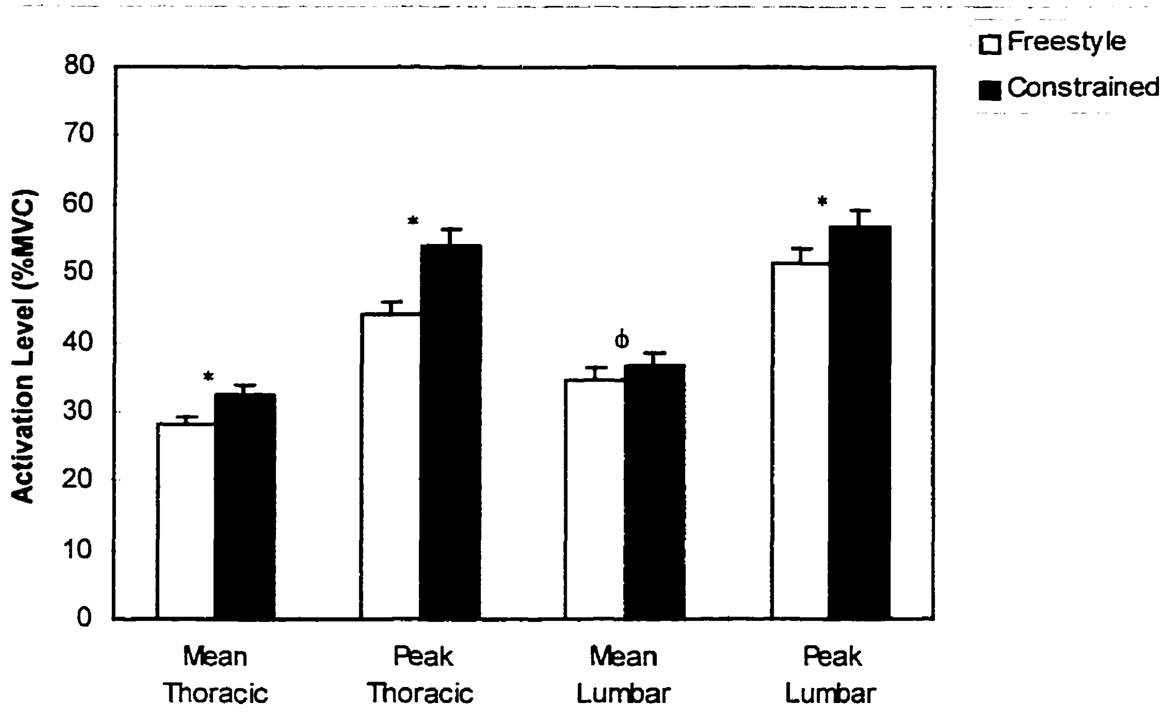

Figure 9. Mean and peak activation levels of both TES and LES muscle groups for freestyle and constrained conditions. Data are pooled across *sex* and *direction* (n = 44). (* = significant *lifting condition* effect at $p < 0.001$, ϕ = $p < 0.05$).

Direction Effect

Kinematic Variables

Mean peak pelvis, trunk, and knee flexion angles, and $\theta_{\%LUM}$ are presented for the lift and lower in Figure 10. Peak pelvic flexion for the lift was significantly less than that for the lower. In contrast, peak knee flexion and $\theta_{\%LUM}$ were greater for the lift than for the lower. No *direction* effect existed for peak trunk flexion angle.

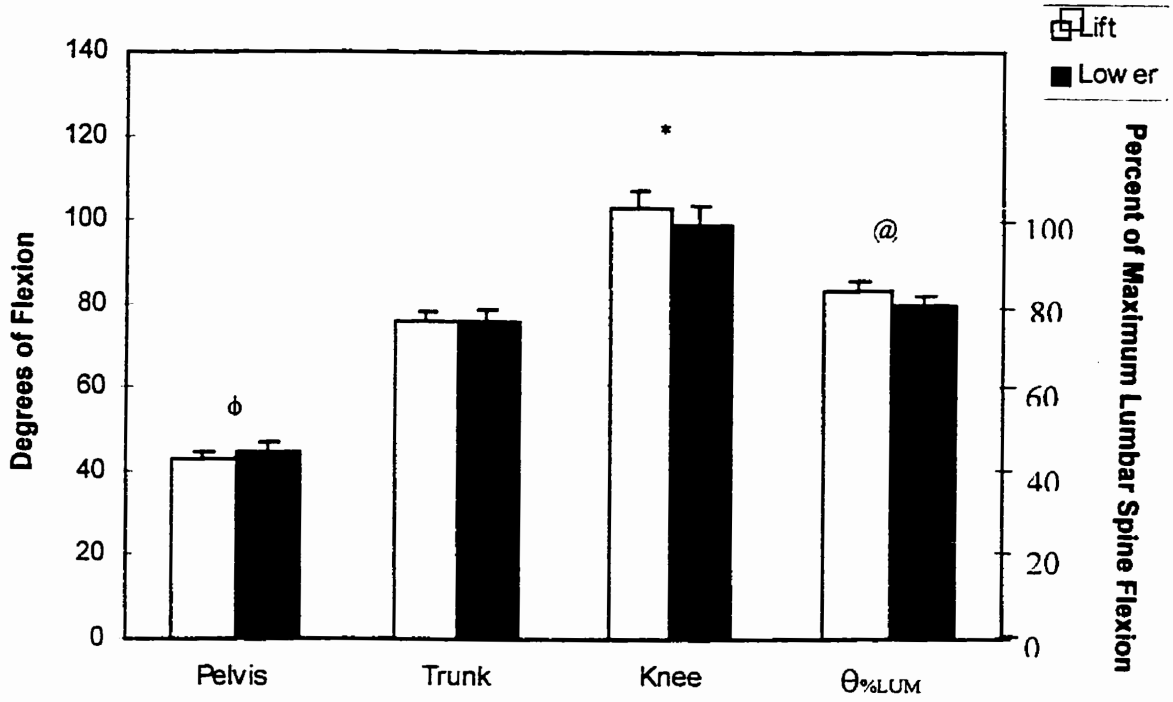

Figure 10. Mean peak Pelvis, Trunk, and Knee flexion angles, and $\theta_{\%LUM}$ for lift and lower directions. Data are pooled across *sex* and *lifting condition* (n = 44). (* = significant *direction* effect at $p < 0.001$, @ = $p < 0.01$, ϕ = $p < 0.05$).

EMG Variables

Mean and peak activation levels of both TES and LES muscle groups are shown for the lift and lower in Figure 11. A consistent trend is seen for all four variables in that within each one, activation levels were all significantly greater during the lift than for the lower.

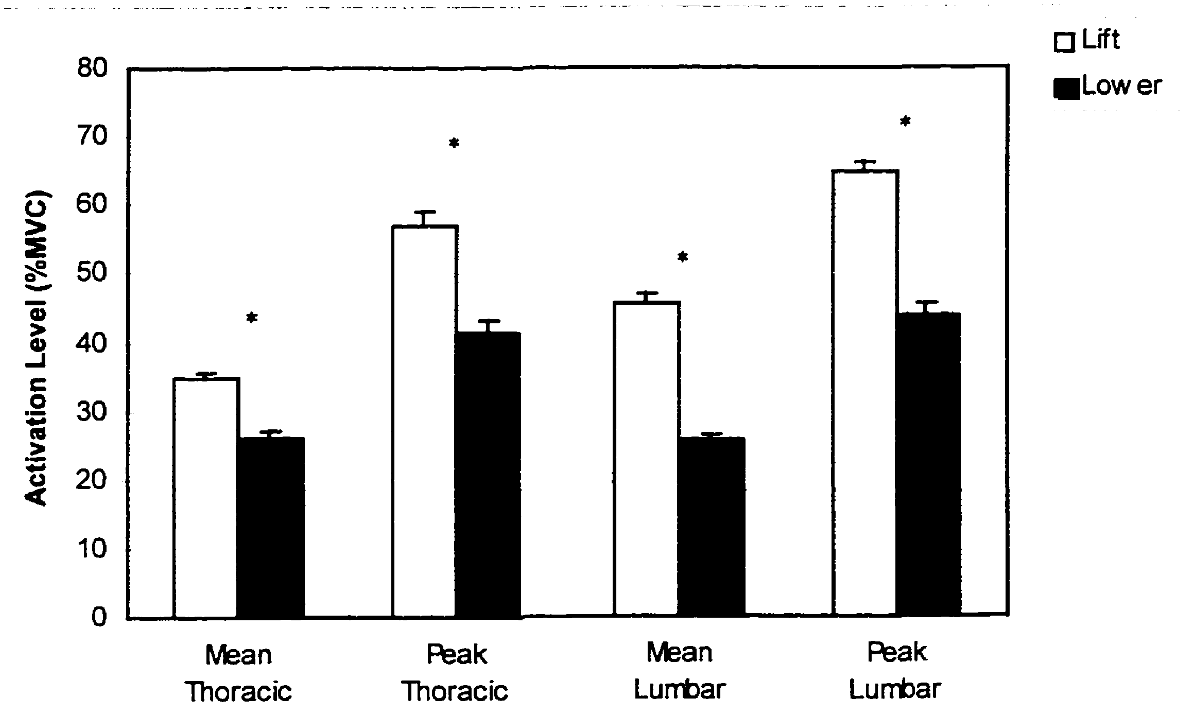

Figure 11. Mean and peak activation levels of both TES and LES muscle groups for the lift and lower directions. Data are pooled across *sex* and *lifting condition* (n = 44). All four variables exhibited a *direction* effect at *$p < 0.001$.

Interaction Effects

Interaction effects are illustrated in Appendix C.

Angular Displacement Time-Histories

Figures 12 through 15 are graphic illustrations of trunk, lumbar spine, knee, and pelvis angular displacements throughout the duration of each of four *lifting condition* x *direction* combinations. Inspection of Figures 12 through 15, several important trends are notable from the angular displacement time-histories that are not seen in the statistical analyses. A very important aspect to note of these four figures is the difference in the pelvis angular displacement curves between freestyle and constrained conditions (Figure 15). For the constrained condition, the pelvis actively extended for the entire duration of the lift, and flexed for the entire duration of the lower. However, for the freestyle condition, there existed intervals during the lift and lower in which pelvic angular displacement was opposite to that observed during the constrained condition. The freestyle lift began with a period of time during which the pelvis was flexing, until it began to actively extend in aiding the trunk to the erect position. For the freestyle lower, the pelvis began by actively flexing, but then proceeded to extend for the latter part of the trial. These distinct time intervals of pelvic angular displacements observed during the freestyle lift and lower will be referred to as "pelvic deviations".

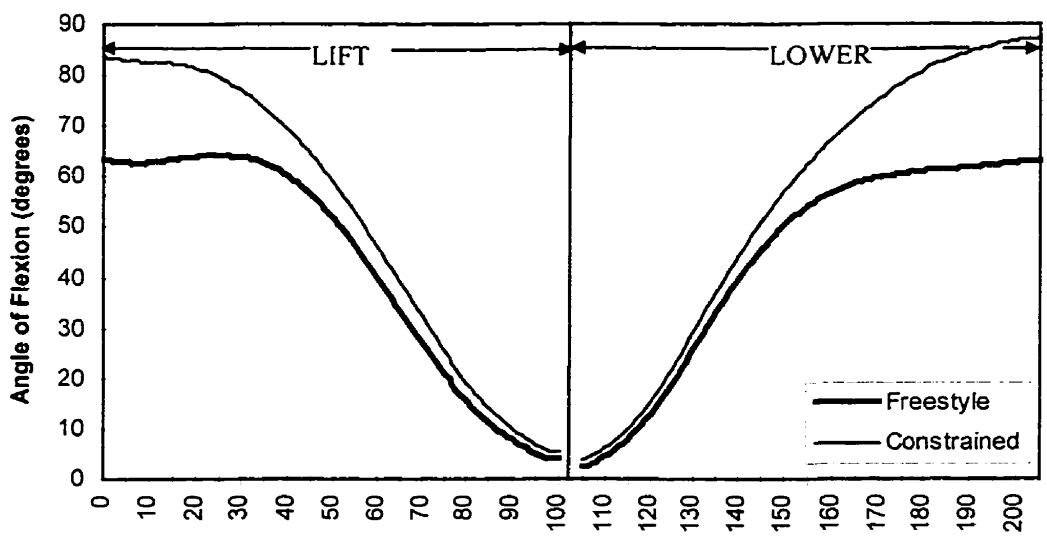

Figure 12. Trunk angular displacement throughout the lift and lower for the freestyle and constrained conditions. Data are pooled across *sex* (n = 22).

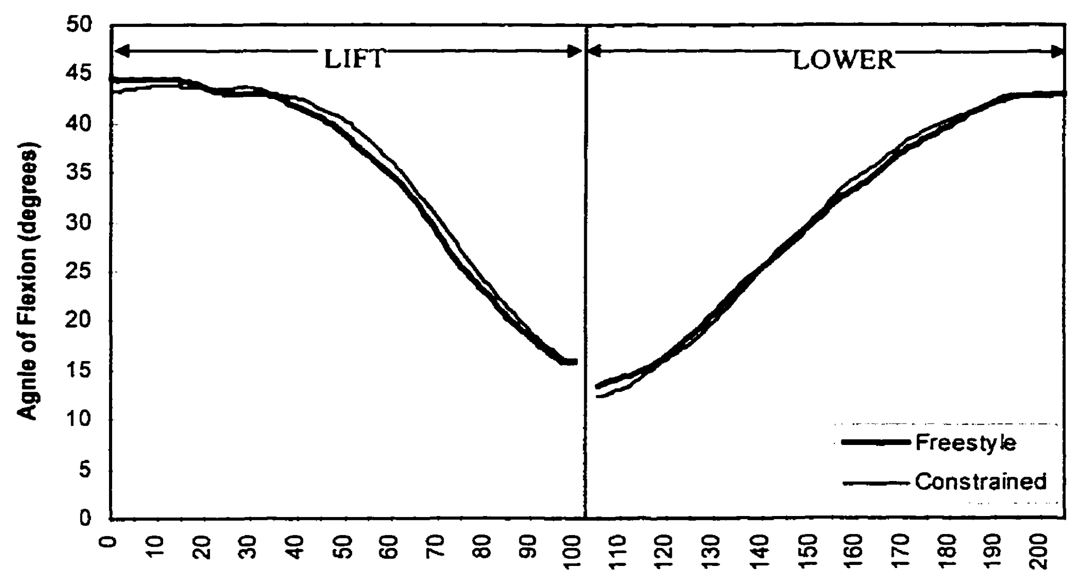

Figure 13. Lumbar Spine angular displacement throughout the lift and lower for the freestyle and constrained conditions. Data are pooled across *sex* (n = 22).

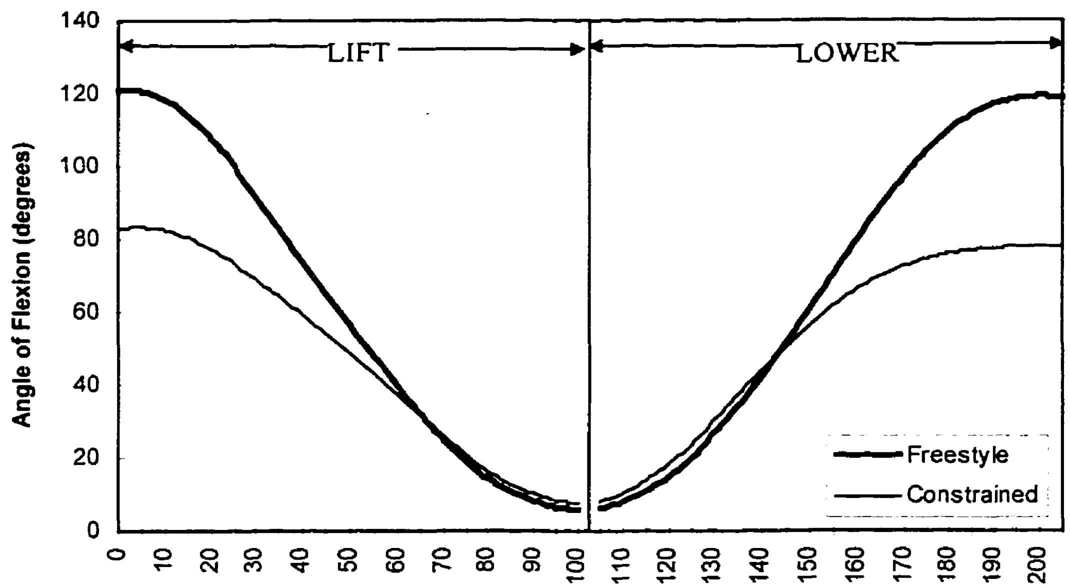

Figure 14. Knee angular displacement throughout the lift and lower for the freestyle and constrained conditions. Data are pooled across *sex* (n = 22).

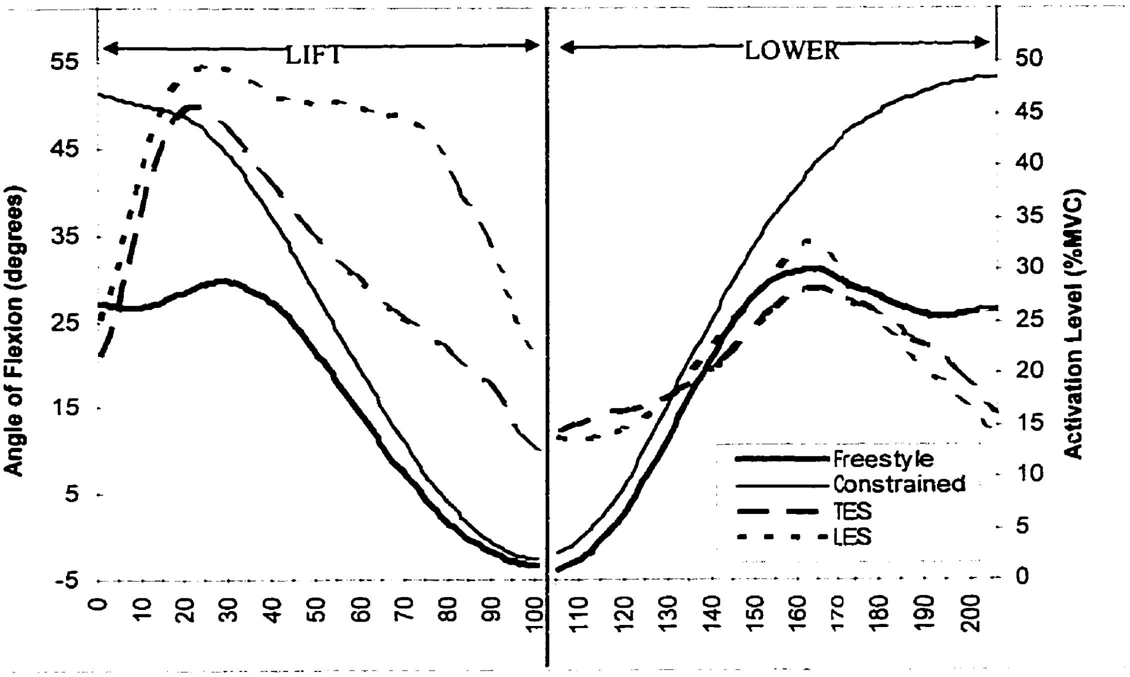

Figure 15. Pelvis angular displacement throughout the lift and lower, for freestyle and constrained conditions, and TES and LES activation levels for the freestyle lift and lower. Pelvis angular displacement is expressed in degrees of flexion (negative values indicate extension past calculated bias angles during standing). TES and LES activation levels are expressed as %MVC. Data are pooled across *sex* (n = 22).

Pelvic Deviations

Figure 15 is a graphical illustration of the timing of TES and LES activation levels along with pelvic angular displacement throughout the lift and lower for both freestyle and constrained conditions. It shows that during approximately the first 25% of the freestyle lift duration, both the TES and LES activation levels ascended to reach their peak. Within this first quarter of the trial, it was at about the time when both muscle groups reached 35% MVC that the pelvis began to flex. Pelvic flexion ceased and extension began as both TES and LES activation levels proceeded to descend from their respective peaks. For the freestyle lower, the pelvis was flexing for the first 58% of duration. It was also during this time frame that the TES and LES activation levels ascended to reach their peak levels. From approximately 58 to 100% of the trial duration, both the TES and LES activation levels descended from their peaks. Furthermore, it was from roughly 58 - 90% of the freestyle lower that the pelvis began to extend.

Pelvis - Lumbar Spine (P-S) Hystereses

Angular displacements of the pelvis versus that of the lumbar spine are shown in Figures 16 through 19. Each figure is a hysteresis plot, illustrating a comparison between two P-S plots of the four *condition* x *direction* combinations. Previously described pelvic deviations at the beginning of the freestyle lift and end of the freestyle lower are also visible in Figure 16. Note that at the beginning of the lift, the pelvic flexion caused the lumbar spine to extend, and conversely at the end of the lower, the pelvic extension put the lumbar spine into further degrees of flexion. As previously seen in the angular displacement time-histories, these pelvic deviations did not exist in the constrained condition (Figure 17). Figures 16 and 17 also illustrate that at a given pelvic flexion angle, lifting was accomplished with a greater degree of lumbar spinal flexion than the lower. Within each condition, it is also visible that lumbar spine flexion angles are similar for the start of the lift and end of the lower, and for the end of the lift and start of the lower. This is also seen for the pelvic flexion angles.

Figures 18 and 19 illustrate the greater peak pelvic flexion angles achieved for the constrained condition versus the freestyle condition. They also show that, although the freestyle lift and lower have the pelvic deviations, peak lumbar spine flexion angles remained similar to those of the constrained condition.

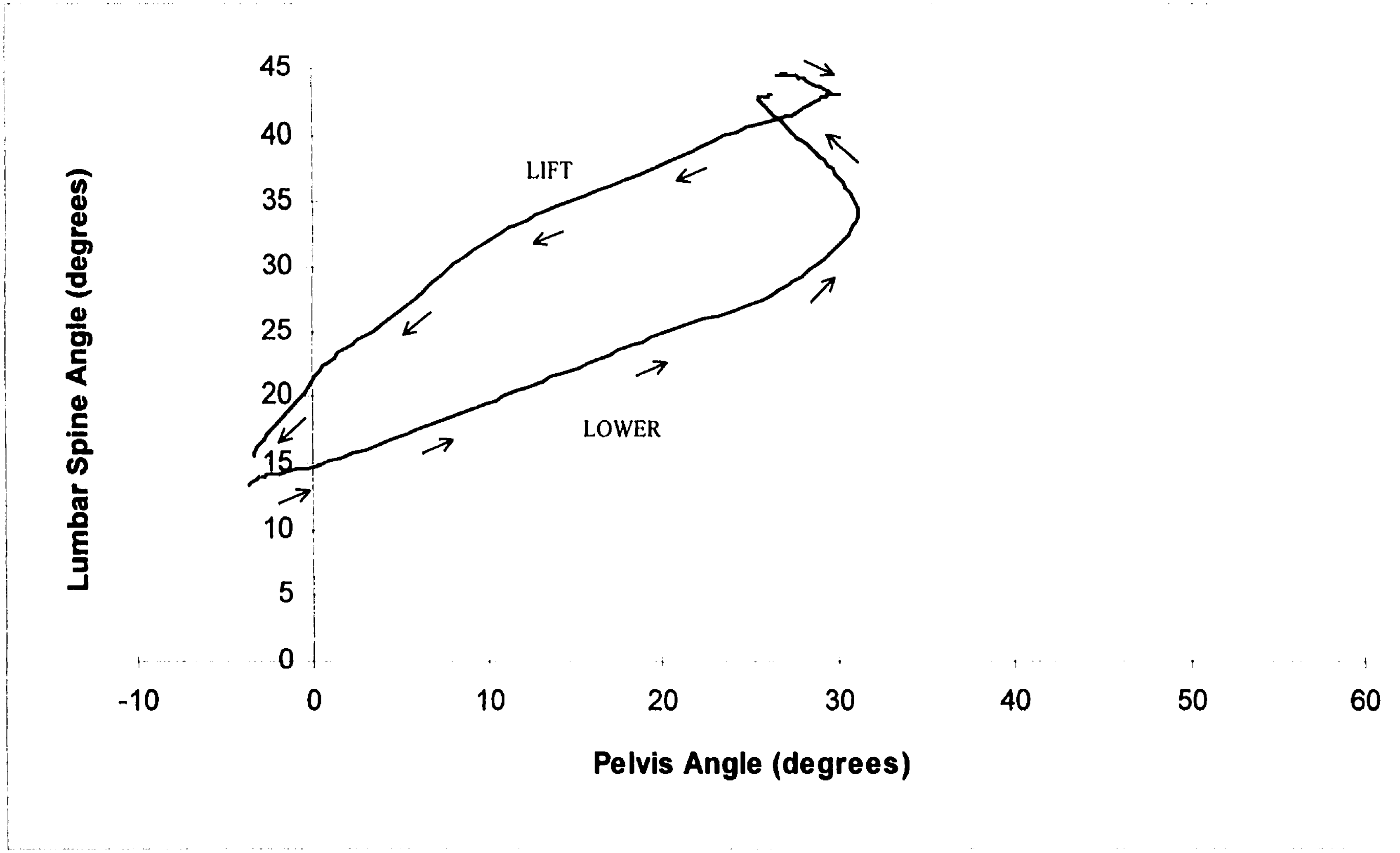

Figure 16. Pelvis - Lumbar Spine hysteresis plot of freestyle lift vs. freestyle lower. Increasing positive angles indicate larger flexion angles. Negative angles represent extension past the calculated bias angles during standing. The lift starts at larger flexion angles and ends at smaller flexion angles (erect stance). The lower follows opposite to the lift. Data are pooled across *sex* (n = 22).

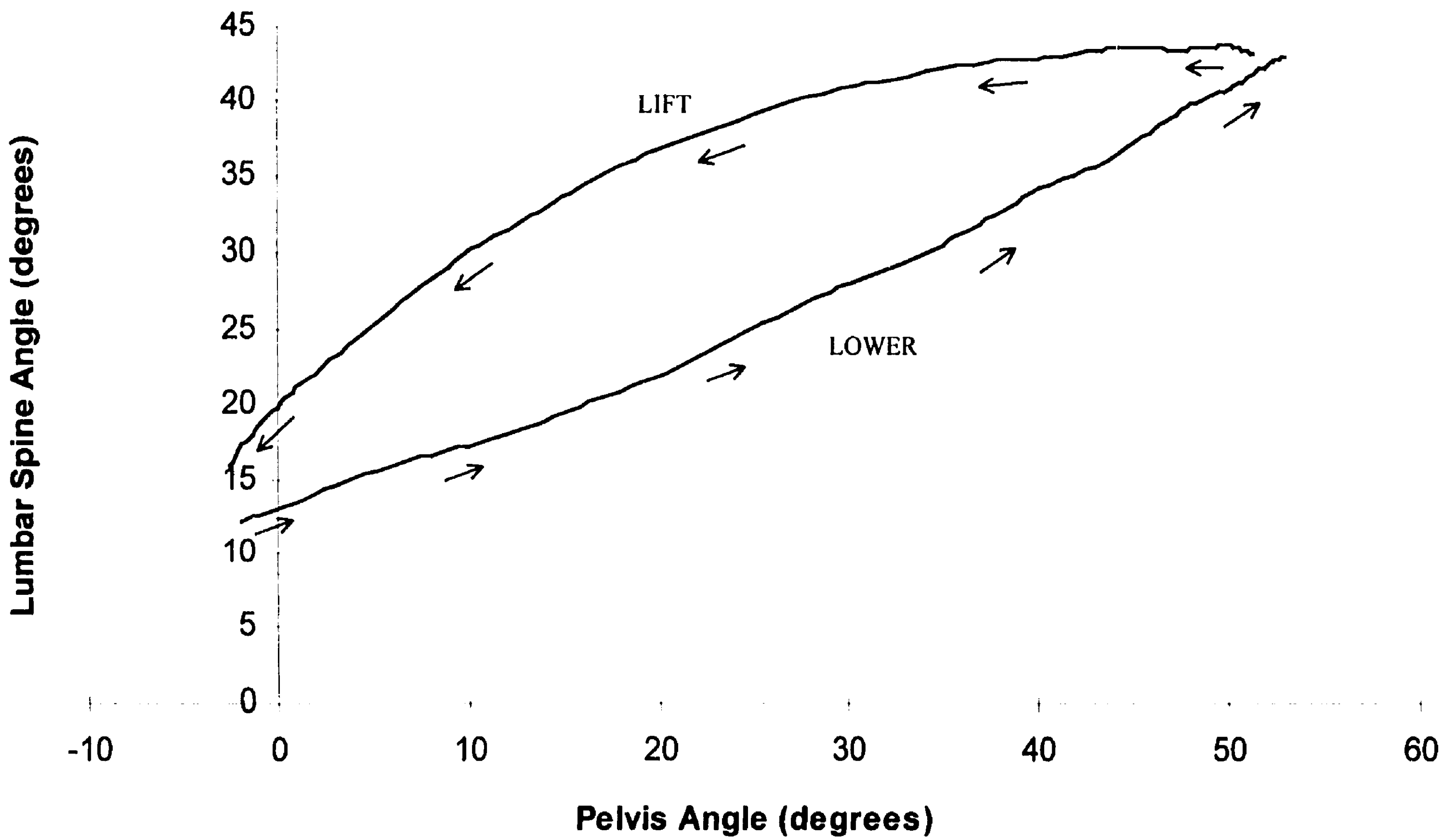

Figure 17. Pelvis - Lumbar Spine hysteresis plot of constrained lift vs. constrained lower. Increasing positive angles indicate larger flexion angles. Negative angles represent extension past the calculated bias angles during standing. The lift starts at larger flexion angles and ends at smaller flexion angles (erect stance). The lower follows opposite to the lift. Data are pooled across *sex* (n = 22).

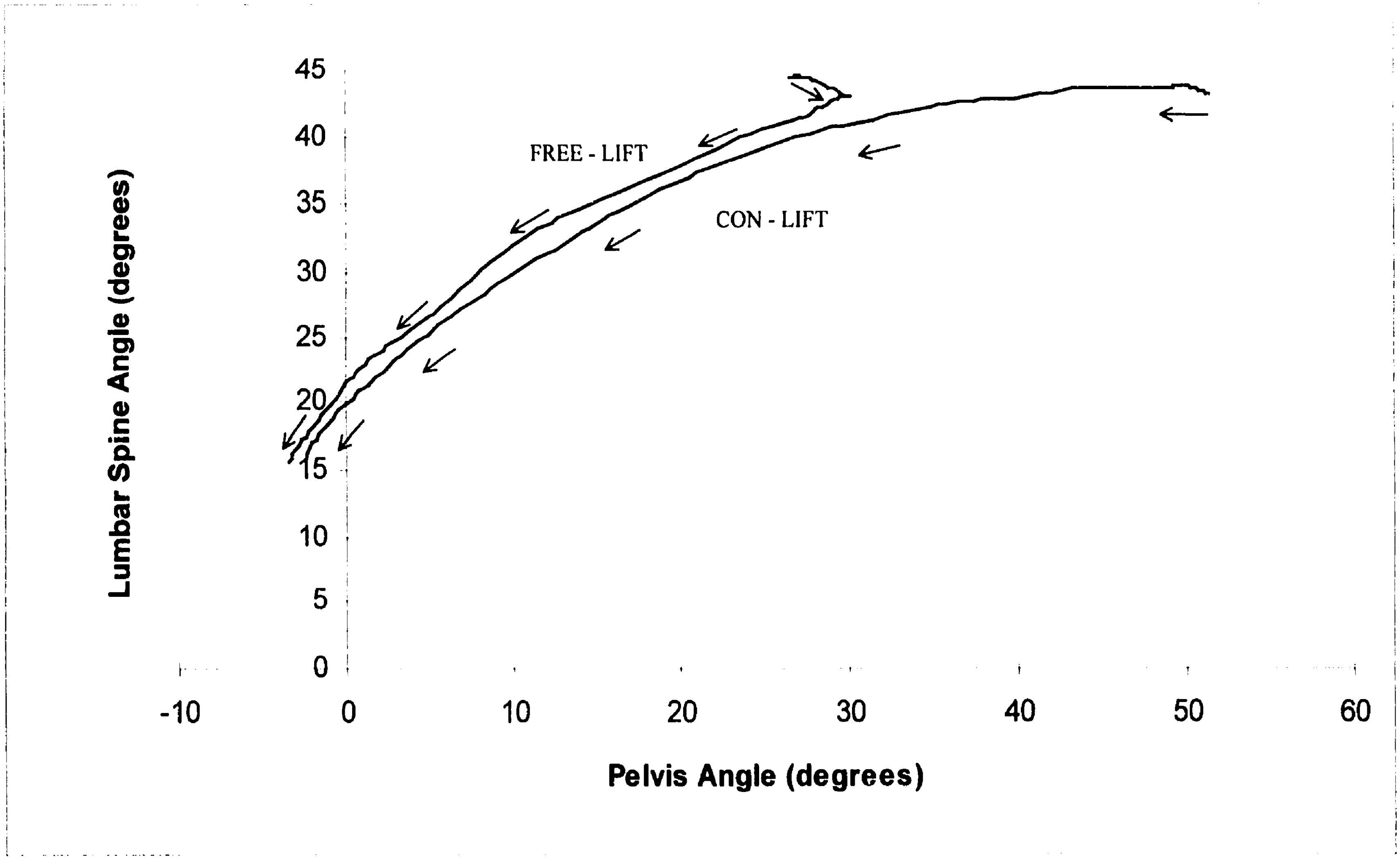

Figure 18. Pelvis - Lumbar Spine hysteresis plot of freestyle lift (FREE - LIFT) vs. constrained lift (CON - LIFT). Increasing positive angles indicate larger flexion angles. Negative angles represent extension past the calculated bias angles during standing. The lifts start at larger flexion angles and end at smaller flexion angles (erect stance). Data are pooled across *sex* (n = 22).

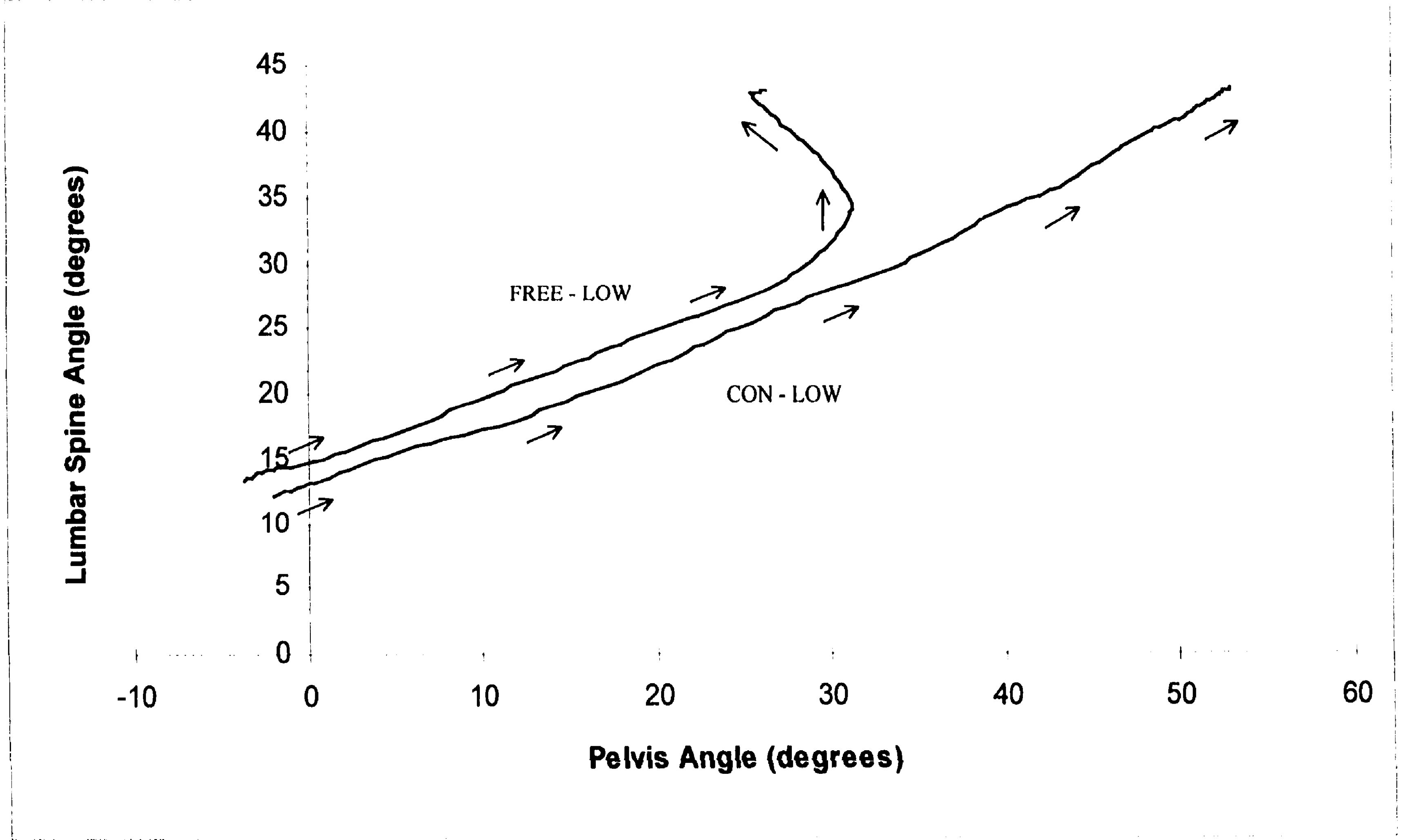

Figure 19. Pelvis - Lumbar Spine hysteresis plot of freestyle lower (FREE - LOW) vs. constrained lower (CON - LOW). Increasing positive angles indicate larger flexion angles. Negative angles represent extension past the calculated bias angles during standing. The lowers start at smaller flexion angles (erect stance) and end at larger flexion angles. Data are pooled across *sex* (n = 22).

Phase Angles

Figures 20 to 23 are graphical representations of the phase angle between the lumbar spine and pelvis for all four *lifting condition* x *direction* combinations. The phase angle between these two structures provides additional information to the P-S hysteresis plots as it is more sensitive to relative changes in pelvis versus spine movement velocity. While both types of plots are sensitive to changes in pelvis versus spine displacement, the phase angle plots more accurately illustrate the changes in velocity between the two structures. Hence, the phase angle plots illustrate the difference in the rate of movement between the pelvis and lumbar spine. While lifting the load (Figures 20 and 22), positive phase represents the rate of pelvic extension exceeding that of the lumbar spine. Negative phase represents the rate of lumbar spine extension exceeding that of the pelvis. While lowering the load (Figures 21 and 23), positive phase represents lumbar spine flexion exceeding that of the pelvis. Negative phase represents flexion of the pelvis exceeding that of the lumbar spine. Figures 20 and 21 illustrate the presence of the pelvic deviations in the freestyle condition. Figure 20 shows that the phase angle during the initial 30% of the freestyle lift was negative due to flexion of the pelvis, subsequently causing the lumbar spine to extend. Figure 21 illustrates a positive phase angle from 40 to 90% of the freestyle lower due to extension of the pelvis which caused the lumbar spine to flex. Figure 23 indicates that the constrained lower was performed with very little phase difference between the lumbar spine and the pelvis.

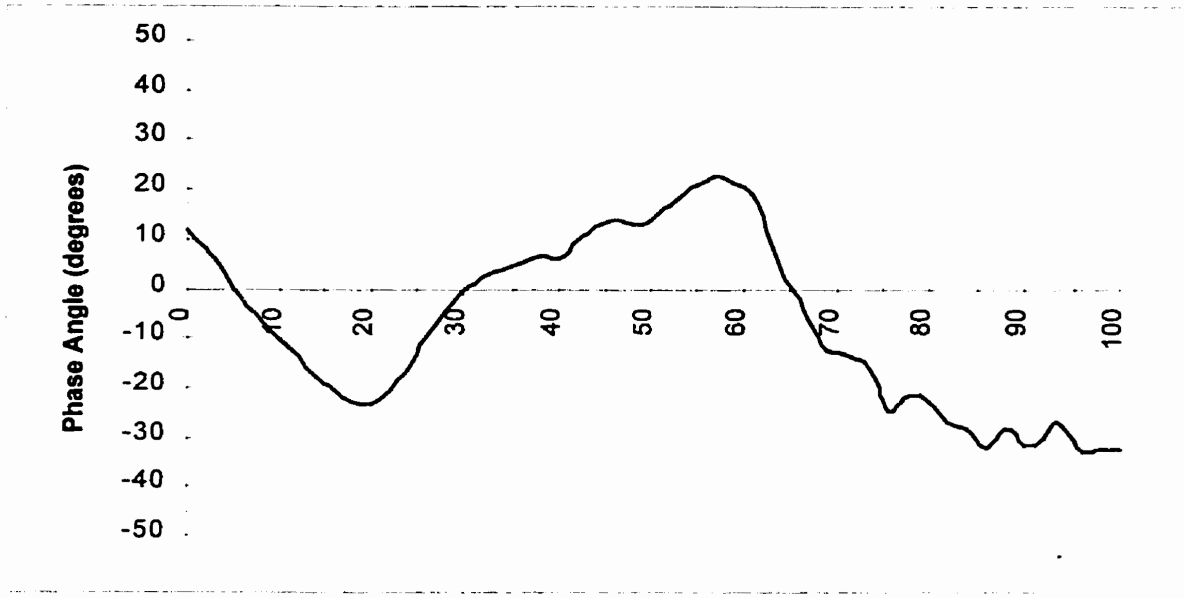

Figure 20. Phase Angle time-history between the Lumbar Spine and Pelvis, for the freestyle lift. Positive phase angle represents the rate of pelvic extension exceeding that of the lumbar spine. Negative phase angle represents the rate of lumbar spine extension exceeding that of the pelvis. Data are pooled across *sex* (n = 22).

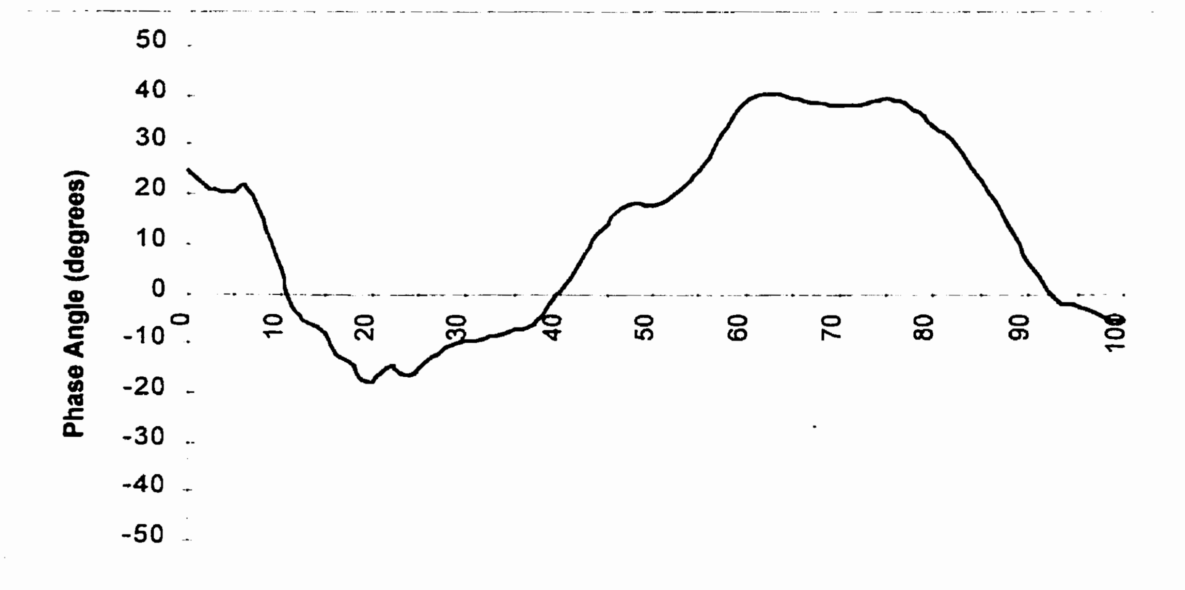

Figure 21. Phase Angle time-history between the Lumbar Spine and Pelvis, for the freestyle lower. Positive phase angle represents the rate of lumbar spine flexion exceeding that of the pelvis. Negative phase angle indicates the rate of pelvic flexion exceeding that of the lumbar spine. Data are pooled across *sex* (n = 22).

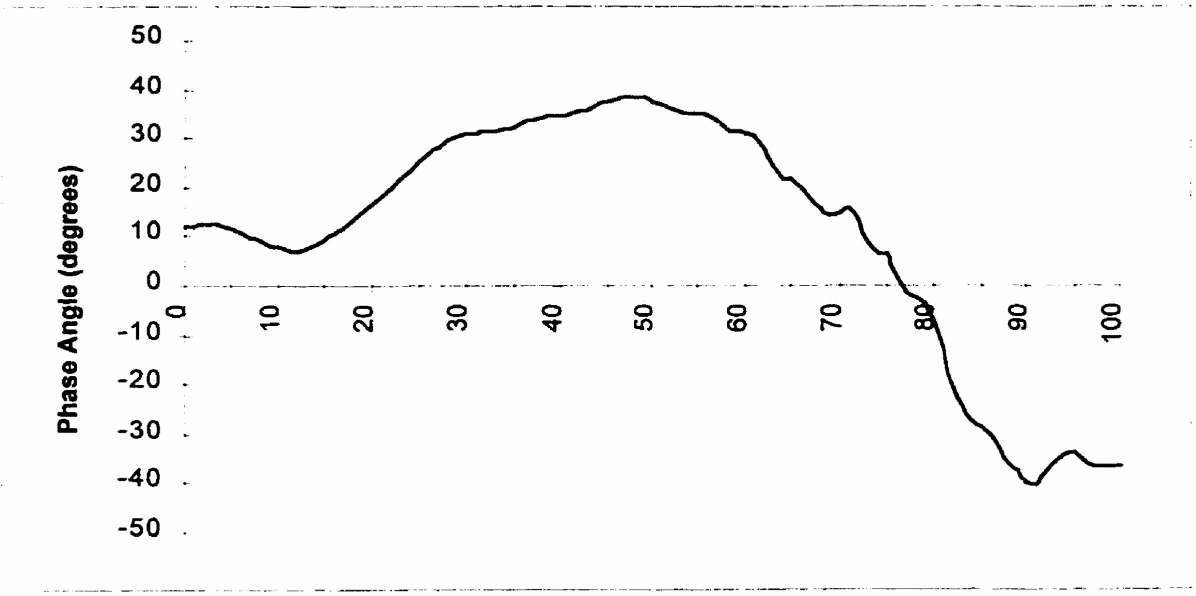

Figure 22. Phase Angle time-history between the Lumbar Spine and Pelvis, for the constrained lift. Positive phase angle represents the rate of pelvic extension exceeding that of the lumbar spine. Negative phase angle represents the rate of lumbar spine extension exceeding that of the pelvis. Data are pooled across *sex* (n = 22).

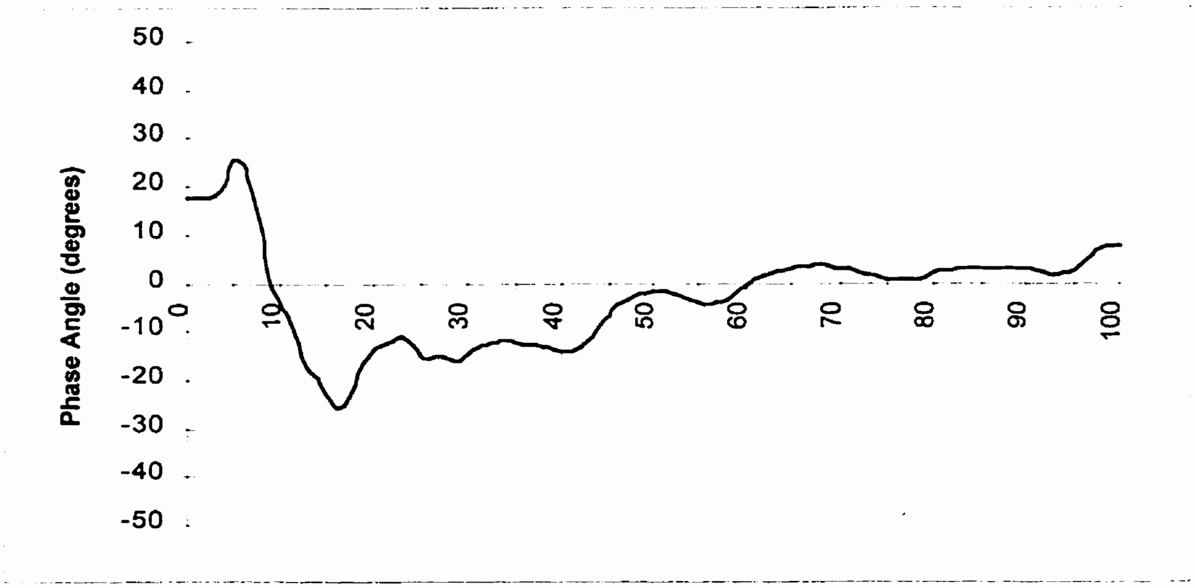

Figure 23. Phase Angle time-history between the Lumbar Spine and Pelvis, for the constrained lower. Positive phase angle represents the rate of lumbar spine flexion exceeding that of the pelvis. Negative phase angle indicates the rate of pelvic flexion exceeding that of the lumbar spine. Data are pooled across *sex* (n = 22).

H-V Hystereses

Horizontal distance of the load from the ankle (H) versus vertical height of the load (V) are shown in Figures 24 through 26. Each figure is a hysteresis plot, illustrating a comparison between two H-V plots. Figures 24 and 25 illustrate that pooled across sex, the load was lifted and lowered at a greater peak H for the constrained condition than the freestyle condition. Figure 26 illustrates that within the constrained condition, the peak H was greater for the lower than the lift for both males and females. No *direction* differences were found between peak H within the freestyle condition for either sex.

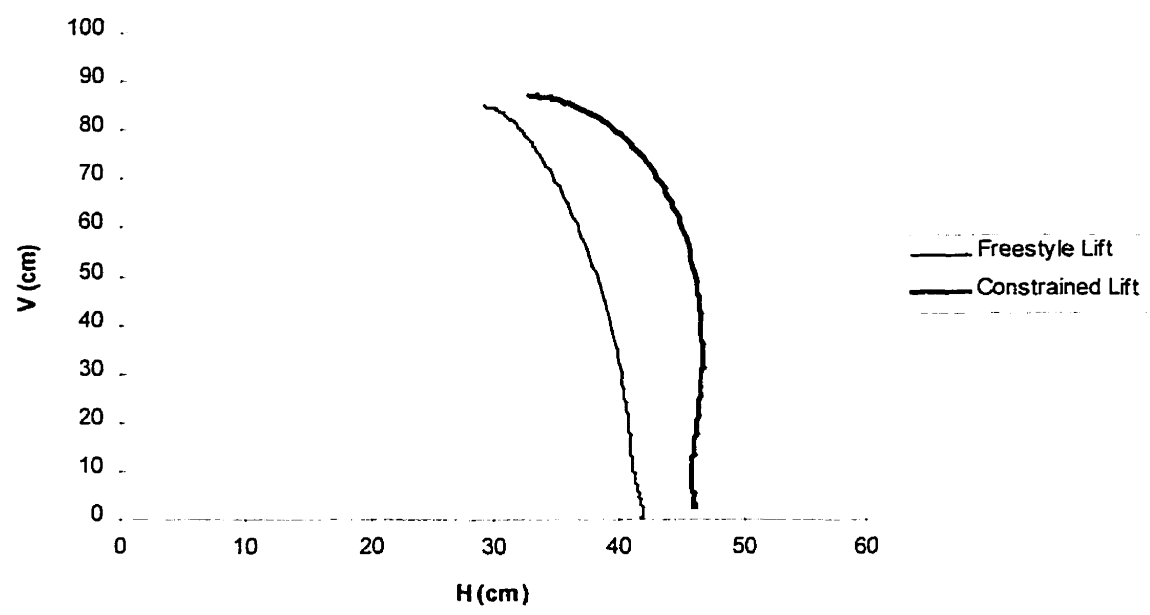

Figure 24. Horizontal vs. Vertical displacement of hand held load from the ankle for freestyle and constrained lift. The lift began with the load on the ground, with vertical displacements of approximately zero. Data are pooled across *sex* (n = 22).

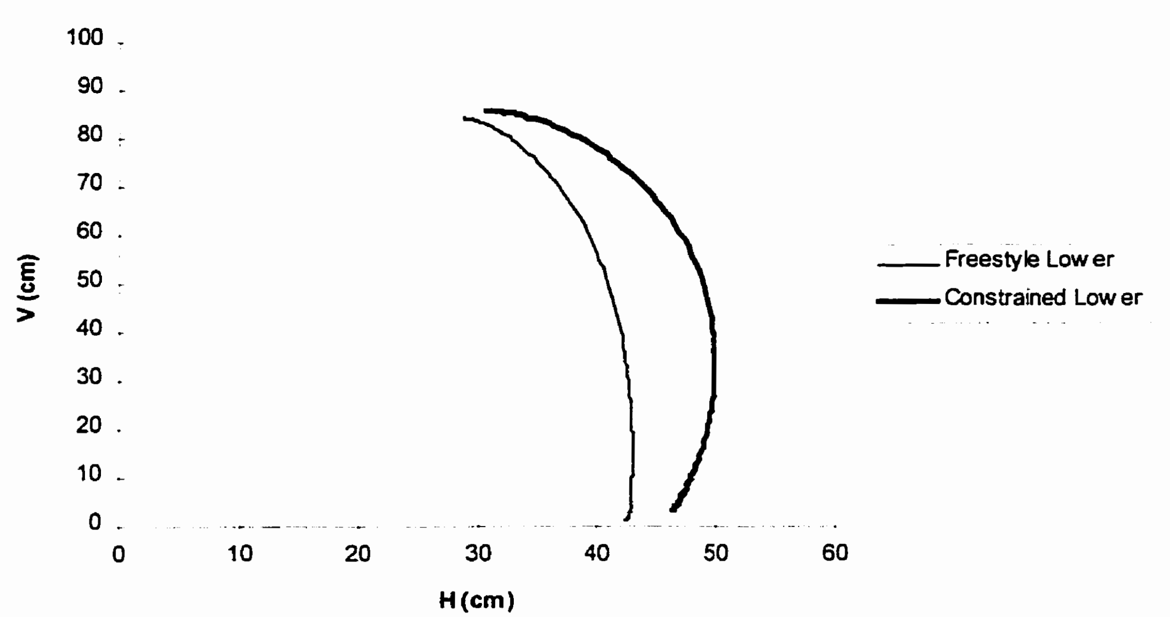

Figure 25. Horizontal vs. Vertical displacement of hand held load from the ankle for freestyle and constrained lower. The lower began with the load being held up from the ground during erect stance. Data are pooled across *sex* (n = 22).

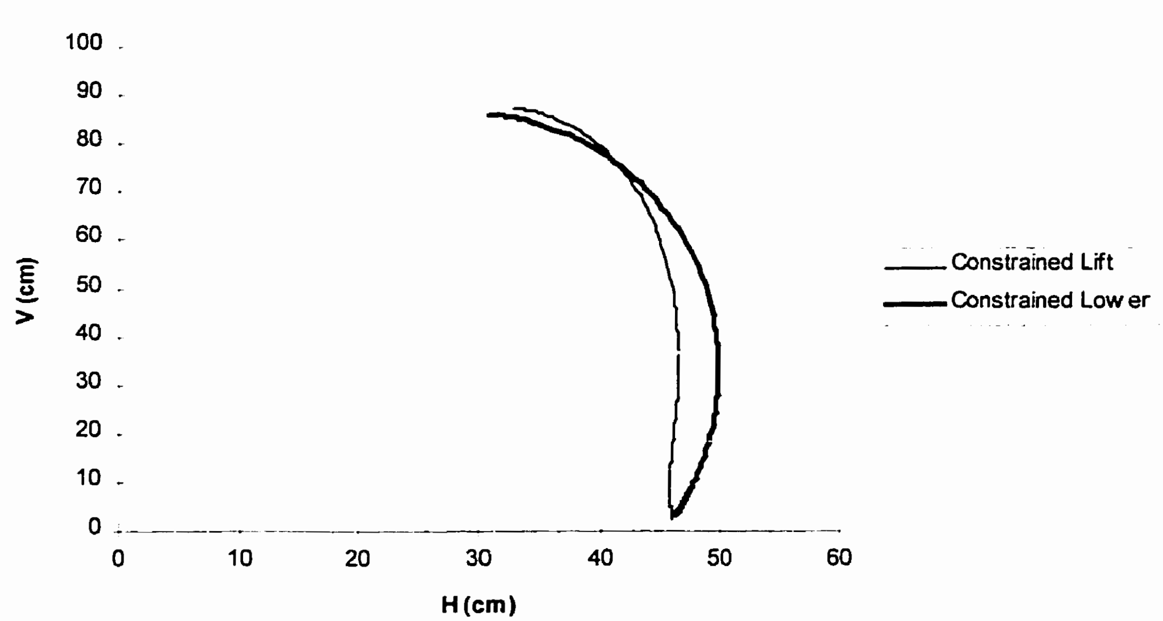

Figure 26. Horizontal vs. Vertical displacement of hand held load from the ankle for constrained lift and lower. The lift began with the load on the ground, with vertical displacements of approximately zero. The lower began with the load being held up from the ground during erect stance. Data are pooled across *sex* (n = 22).

Flexion-Relaxation Phenomenon

There was no period of electrical silence of the LES muscle group under any condition, and hence no evidence of the flexion-relaxation phenomenon.

Chapter 5

DISCUSSION

The purposes of this study were four fold: (a) to determine whether sagittal plane movement of the lumbar spine and pelvis occur synchronously, (b) to investigate whether sex differences exist in lumbar spine - pelvic kinematics, (c) to determine whether peak lumbar spine flexion differs between lifting and lowering, and (d) to observe the relative contributions of the lumbar spine and pelvis to increases in trunk flexion. These were all investigated under conditions of manual lifting and lowering. Kinematic variables under observation were mean peak trunk, knee, lumbar spine, and pelvic flexion angles. All flexion angles were expressed in degrees, except for the lumbar spine. Peak lumbar spine flexion angles were expressed as a percent of maximum flexion. The time-histories of lumbar spine and pelvis sagittal plane flexion/extension were also collected for each condition for the determination of their pattern of movement. Kinetic variables were mean and peak thoracic and lumbar erector spinae EMG levels.

Main Findings

Results of this study indicate a synchronous pattern of movement between the lumbar spine and pelvis for all conditions of loaded sagittal plane trunk motion, except the constrained lift (Figures 16 and 17). Trunk extension is initiated by posterior rotation of the pelvis at the start of a lift when a constraint such as a wall is placed in front of a person (Figure 17). Hence, the pattern of sagittal plane motion between the lumbar spine and pelvis is sequential under this type of condition. This study is unique in that it is the only one to the author's knowledge that has monitored lumbar spine and pelvis kinematics under a realistic constraint condition that has direct applications to the workplace. Intuitively, greater flexion of the trunk has always been attributed to increases in lumbar spine flexion. However, this study has shown that the greater trunk flexion associated with a constrained condition is primarily a result of further rotation of the pelvis (Figure 8). Despite the absence of any change in lumbar spine flexion between conditions, it is still at a greater risk for injury for the constrained condition, compared to the freestyle condition, as the load is held further away from the lumbar spine's axis of flexion (Figures 24 and 25). The difference in the magnitude of lumbar spine flexion between lifting and lowering is small, indicating very little difference in low back injury risk between the two directions of load handling. No sex differences exist in lumbar spine - pelvic kinematics during manual materials handling. This is supported by the same trend observed in erector spinae muscular activation levels.

Hypotheses

Synchronic Movement of the Lumbar Spine and Pelvis

It was hypothesized that movement of the lumbar spine and pelvis would be synchronous during sagittal plane trunk motion while handling loads. Very limited research has been conducted on the lumbar spine - pelvic pattern of movement during loaded conditions.

Calculations of relative phase angles by Burgess-Limerick *et al.* (1992) have attempted to mathematically quantify the pattern of movement between these two structures. This supports the conclusions of Potvin *et al.* (1991) that the lumbar spine and pelvis were moving in a synchronous pattern during sagittal plane lifting.

This hypothesis was accepted for all conditions of this study, except the constrained lift. Winstein and Garfinkel (1989) have illustrated plots of perfectly synchronous movement between two structures to have a slope magnitude of 1. On the spine-pelvis hysteresis plots (Figures 16 to 19), perfectly synchronous movement of the two structures would be identified as a plot with a linear relationship, but the slope of this line would depend on the magnitude of maximum possible flexion of each structure. The same pattern type of movement would be indicated on the phase angle plots (Figures 20 to 23) as a horizontal line (slope = 0) along the Y axis value of zero. This would indicate the absence of any phase angle between the movement of the two structures. Analysis of both types of plots revealed that sagittal movement of the lumbar spine and pelvis did not occur in a perfectly synchronous pattern. However, for three of four conditions, they were in motion at the same time, with one at a greater rate than the other. Trunk extension for the constrained lift was, however, initiated almost solely by posterior rotation of the pelvis, until the lumbar spine also began to extend (Figure 17).

Sex Effects

It was hypothesized that males would accomplish the tasks with greater peak flexion of the lumbar spine, while females would do so with greater peak pelvic flexion.

Preliminary data collected by Ursulak and Potvin (1995) during three variations of lifting led to this hypothesis.

This hypothesis was not accepted. No sex differences were detected for either of these two variables or for the mean and peak TES and LES EMG levels.

Lift vs. Lower Effects

It was hypothesized that peak lumbar spine flexion would be greater at the end of the lower when compared to beginning of the lift. The basis for this hypothesis is that spine forward flexion is a result of the force of gravity acting on the upper body and the hand held load, with the back muscles contracting eccentrically to control the rate at which this occurs (Bogduk and Twomey, 1991; Oliver and Middleditch, 1991). Thus, an individual is actually being pulled down to the ground under the force of their upper body and load, when lowering. By this reasoning, the lumbar spine should flex more at the end of lowering the load, as opposed to the start of lifting.

This hypothesis was not accepted. Subjects flexed the lumbar spine 3.8% more at the start of the lift than at the end of the lowering tasks (Figure 10). Mean and peak TES and LES EMG levels were also significantly greater for the lift than the lower (Figure 11).

Freestyle vs. Constrained Lifting Effects

The wall was placed in front of the subjects to restrict flexion at the knees. Thus, it was hypothesized that there would be an increase in mean peak trunk flexion to compensate for this restriction. Upon comparing squat versus stoop lifting, substantial increases in trunk flexion have been observed due to the more extended knee posture during the stoop lift (Park and Chaffin, 1974; Garg and Herrin, 1979; Potvin *et al.*, 1991).

This hypothesis was accepted. The constrained condition significantly reduced the amount of flexion at the knees and, as a result, mean peak trunk flexion increased by 20.1° over the freestyle condition (Figure 8).

It was also hypothesized that the increase in peak trunk flexion for the constrained condition would be partly due to an increase in lumbar spine flexion ($\theta_{\%LUM}$). Furthermore, it was proposed that constrained $\theta_{\%LUM}$ would reach a certain magnitude that would elicit the flexion-relaxation phenomenon (FRP). Increases in lumbar spine flexion have been shown to contribute to the requirement of greater trunk flexion angles while handling loads (Potvin *et al.*, 1991; De Looze *et al.*, 1993; Ursulak and Potvin, 1994).

These hypotheses were not accepted. Results indicated that $\theta_{\%LUM}$ remained constant between freestyle and constrained conditions (Figure 8). The LES EMG signal did not show any period of silence, therefore signifying the absence of the FRP.

Finally, it was hypothesized that although $\theta_{\%LUM}$ might increase somewhat, the main contribution to the increase in mean peak trunk flexion for the constrained condition would come from the pelvis. Results from previous lifting studies have shown that the requirement of greater trunk flexion angles have mainly been attributed to increases in pelvic flexion (Potvin *et al.*, 1991; Ursulak and Potvin, 1994).

This hypothesis was accepted. Mean peak pelvic flexion increased by 18.4° from the freestyle to constrained conditions (Figure 8). Hence, the increase in mean peak pelvic flexion was not only the main contributor, but was the sole source of the augmentation in mean peak trunk flexion for the constrained over the freestyle condition.

Limitations

There are several limitations that should be discussed with respect to this study. Firstly, the fact remains that this study was restricted to symmetrical, two-handed load handling in the sagittal plane. The conclusions based on the results of this study are therefore most applicable to industrial conditions that are similar to the ones performed in the laboratory. It must be noted that this is not the only scenario in which manual materials handling occurs in industry. There exists a wide variety of scenarios that involve asymmetric loading, axial twist, or lateral bending that would alter the relationships found in this study.

It is important to note that the orientation of the lumbar spine's lordotic posture during upright stance may have been a source of variability within the data. Standing lumbar lordosis was used in this thesis as a bias angle within the trials. This is a technique commonly employed in kinematic studies monitoring movement of the lumbar spine. Despite this fact, there is the possibility that intersubject differences in erect stance lumbar lordosis may be a source of variability.

Another issue that needs to be addressed is the movement artifact that may have been introduced to the displacement signals by motion of the skin over bony landmarks. Some may argue that this type of kinematic analysis introduces error in tracking the true motion of joints. However, Adams *et al.* (1986) have observed high correlation between skin surface adhered inclinometer measurements and X-ray measurements of varying degrees of lumbar spine flexion (r = 0.91). While inclinometers were not used in the current study, Adams *et al.* (1986) have illustrated that the angle of flexion of the lumbar spine can be accurately measured with a technique that involves adhering measurement devices to the skin overlying the sacrum and the L_1 spinous process.

This study involved the use of electromyography as a representation of the dynamic force generated by the erector spinae musculature. Accurate dynamic EMG - to - force relationships must take into account the change in both muscle length and muscle contraction velocity. EMG signals collected in this study were not corrected for these two variables. Previous studies investigating similar phenomena also did not appear to correct for these factors (Potvin *et al.*, 1991; Ursulak and Potvin, 1995). It was assumed in this study that the incorporation of these modulation factors would not have a substantial effect on the final outcome of data. Asssuming that in upright standing the back extensor muscles are at their optimum position in the force-length curve, then any lengthening of these muscles would move them away from the optimum position on the curve (Winter, 1990). Any electrical activation at such point in time would therefore result in a decrease in force generation from the muscles. At the same time, when taking the force - velocity relationship into account, eccentric contractions of muscle would lead to an increase in force output for any given electrical activation (Winter, 1990). Hence, during forward flexion of the trunk, these two relationships would counteract each other, possibly resulting in a negligible net effect in the electrical signal being collected. During extension of the trunk from a flexed position, the agonistic muscles were returning back to their optimum length for force generation, and were contracting concentrically. With moving in the opposite direction from flexion, the previously mentioned relationships between electrical activity and force generation would be reversed, yet the final outcome would still be an insignificant net effect on the myoelectric signal. These corrections are most important in asymmetrical tasks where there exists left/right differences in muscle length and contraction velocity. These differences in force levels will therefore show force asymmetries on the intervertebral disc.

Another limitation to this study was the subject population. Twenty-two subjects were selected from a university student population. Assumptions were made that the results collected from this group of subjects could represent the North American population of young, healthy adults. However significant the results, caution must be taken in that they are a small sample chosen to represent the average individual in a much larger population.

Synchronic Movement of the Lumbar Spine and Pelvis

Previous studies have presented conflicting data regarding the pattern of movement between the lumbar spine and pelvis. Some reported that the lumbar spine and pelvis moved synchronously during sagittal plane lifting tasks (Potvin *et al.*, 1991; Burgess-Limerick *et al.*, 1992). Others claimed that lifting was performed with a sequential pattern of movement between these two structures (Farfan, 1975; Farfan and Lamy, 1977; Gracovetsky *et al.*, 1977; Anderson *et al.*, 1985; Nelson *et al.*, 1995; Toussaint *et al.*, 1995). The limited number of studies that have analyzed the pattern of motion between the lumbar spine and pelvis while lowering loads have observed it to be synchronous (Nelson *et al.*, 1995; Toussaint *et al.*, 1995).

Conclusions based on studies that mathematically modelled the lumbar spine during lifting have little relevance to this thesis in that pelvic mobility was not monitored (Farfan and Lamy, 1977; Gracovetsky *et al.*, 1977). Their reasoning of a sequential pattern of movement is described by the following: at lumbar spine flexion angles greater than 40°, the extensor moment demand exceeds the capabilities of the erector spinae musculature. They believe the posterior spinal ligaments have an effective moment arm to the spine axis of flexion that is longer than the musculature, and would therefore exert a smaller magnitude of compressive force on the lumbar spine for the same moment. Thus, the ligaments would bare the extensor moment until the pelvis extended far enough such that the demands could be met by the erector spinae musculature (Gracovetsky *et al.*, 1977). It has been reported, however, that the grouped action of the posterior ligaments has an equivalent moment arm of 5.8 cm (McGill, 1988). This length is smaller than that of the grouped action of the lumbar erector spinae, as determined with young men at the L_4/L_5 level (5.9 cm) (McGill *et al.*, 1988). It is therefore apparent that no mechanical advantage exists for the ligaments over the erector spinae musculature in the development of the extensor moment. This calls into question the rationale provided by Gracovetsky *et al.* (1977) for sequential movement between the lumbar spine and pelvis during lifting.

Anderson *et al.*, (1985) based their conclusions on a biomechanical model of the lumbosacral joint (L_5/S_1). They did not monitor sagittal plane movement of the lumbar spine or pelvis, but rather predicted L_5/S_1 and sacral rotation using regression equations based on trunk and knee flexion variables. Reliability of these equations are under scrutiny due to high variability: 7.4% and 16.4% standard error for the L_5/S_1 and sacral rotation equations respectively.

While results from three of four conditions in this thesis support those of Potvin *et al.* (1991) and Burgess-Limerick *et al.* (1992), there are limitations to these previous studies. Potvin *et al.* (1991) employed lifting conditions that did not allow their subjects to perform in a manner in which they felt most comfortable. They also based their conclusions on mean contributions of the pelvis and lumbar spine to the total trunk flexion for all conditions. As all values were recorded when peak lumbar spine flexion occurred, the presence of significant pelvic flexion at these instances led the authors to believe that sagittal motion of these two structures was occurring simultaneously. While lumbar spine flexion was plotted against pelvic flexion for one trunk flexion and extension trial, no mean subject population data was reported that illustrated the pattern of movement between these two structures throughout each condition. Burgess-Limerick *et al.*, (1992) employed the calculation of phase angles for the mathematical quantification of the lumbar spine - pelvic pattern of movement. However, they had too few subjects (n = 3) and very noisy data.

Nelson *et al.* (1995) studied lumbar spine and pelvic kinematics during lifting and lowering of loads with a relatively large subject pool (n = 30). They normalized vertical location of the load so that it would cause each subject to flex their spine to induce exactly 90% of their maximum flexion range. Subjects, however, were instructed on lifting technique: keeping the knees and elbows extended during the tasks. It should also be noted that within each subject's trial, lumbar spine and pelvic flexions were normalized to the maximum flexion obtained in that trial. This form of normalization does not have much functional relevance as the absolute magnitude of flexion is unknown.

Results of this thesis were based on conditions that better simulated a more natural displacement of the body as subjects were able to perform the tasks in a manner in which they felt most comfortable. An adequate population sample size was selected (n = 22). Pelvic displacement was based on rotation of a triangular fin on the sacrum, while lumbar spine displacement was calculated as the difference between that of the ribcage and pelvis. This calculation was based on that employed in previous sagittal plane lifting studies (Potvin *et al.*, 1991; Ursulak and Potvin, 1994). Absolute magnitude of lumbar spine flexion was plotted against that of the pelvis for all conditions so that the true pattern of motion could be observed between the two structures. In addition, calculations of phase angles were adopted from Burgess-Limerick *et al.* (1992) in order to monitor the rate of motion difference between the two structures for all conditions.

It has been previously observed that early stages of lifting are coupled with further lumbar spine flexion (Troup, 1977; Nelson *et al.*, 1995). This has been proposed as being the result of pelvic extension being the sole contributor to trunk extension in the initial stages of lifting (Nelson *et al.*, 1995). Hence, following the theory of Gracovetsky *et al.* (1977), this pattern of movement would reduce risk of injury to the lumbar spine as further flexion would increase the stress on the posterior spinal ligaments so that they may bare the extensor moment. The current thesis has shown that even under conditions in which trunk extension is initiated by posterior rotation of the pelvis, no further flexion of the lumbar spine occurs (Figure 17). Movement of the lumbar spine was under muscular control for all conditions of load handling. This was indicated by the absence of the flexion-relaxation phenomenon (FRP) for all conditions. During loaded conditions, occurrence of the FRP takes place upon flexion of the lumbar spine to approximately 96% of it's maximum (Kippers and Parker, 1984). Average lumbar spine flexion in the current thesis did not exceed 81.6% of it's maximum. Potvin *et al.* (1991) observed lumbar spine flexion to reach 84.4% of maximum for stoop lifts and observed no occurrence of the FRP.

The lumbar erector spinae (LES) exert a posterior translational force on the lumbar vertebrae that increases in magnitude as the vertebral level of the lumbar spine increases (Bogduk and Twomey, 1991). Due to the steep orientation of the posterior lumbar spinal ligaments with respect to the mid-disc plane, it has been proposed that they do not have a very prominent role in the development or resistance of translational (shear) forces (Panjabi *et al.*, 1991; Dickey, 1998). Hence, maintenance of lumbar spine flexion to levels at which the moment demands are balanced primarily by the musculature allows the LES to support against anterior shear forces that would otherwise not be supported against if they were inactive.

Sex Effects

Results from the current study indicate that no sex differences exist in lumbar spine - pelvic kinematics during sagittal plane manual materials handling. Dolan *et al.* (1994) have also monitored sagittal plane spinal motion for both males (n = 23) and females (n = 126) during dynamic lifting tasks. Similar to the current thesis, lumbar flexion was expressed as a percent of absolute maximum for squat, freestyle and stoop lifts. No appreciable sex differences were observed in lumbar spine flexion within each lifting technique. Since kinematic results from the current thesis show that no sex differences existed in spine kinematics and trunk extensor EMG, differences in injury statistics may very well be due to conventional thoughts. Males have been observed to be 30% stronger than females of equivalent height, weight, and training (Hayne, 1981). Hence, it may be that males typically occupy jobs that are more physically demanding than those occupied by females. It may be for this reason that male mean LBP claim costs are greater than those for females.

Lift vs. Lower

Few studies have compared lumbar spine kinematics between lifting and lowering of loads. De Looze *et al.* (1993) investigated lift/lower differences in L_5/S_1 joint moments as a result of peak flexion angle and acceleration of the body's center of mass (n = 8). This was conducted under "leg" and "back" bending techniques. They observed that within each technique of bending, no significant lift/lower difference existed in peak L_5/S_1 flexion angle. Kinematic data were averaged over only two trials of eight subjects. Load handling techniques were instructed, and directed at a pace set by a metronome.

Results from the current thesis indicate that peak lumbar spine flexion is greater during lifting than lowering. Subjects were able to perform the tasks at a natural cadence, in a manner that was most comfortable to them. Kinematic data were averaged over 3 trials for all 22 subjects, therefore yielding greater statistical strength than De Looze *et al.* (1993). In addition, sagittal plane motion of the entire lumbar spine was monitored in this study, as opposed to only the L_5/S_1 joint being monitored by De Looze *et al.* (1993).

Marras *et al.* (1993) performed an industrial surveillance study from which maximum sagittal spine angle and maximum load moment were included as two variables that were closely associated with jobs that had high risk for low back disorder (LBD). Current results would therefore indicate that individuals are at a slightly greater risk for LBD during lifting rather than lowering of loads. Progressive levels of lumbar spine flexion have been observed to induce increased strain on the interspinous and supraspinous ligaments (McGill, 1988). These ligaments have been shown to be the first to sprain in cases of lumbar spine hyperflexion (Adams *et al.*, 1980). Lifting conditions would therefore predispose individuals to somewhat greater risk of spraining the passive tissues of the lumbar spine than lowering.

Studies have illustrated greater erector spinae muscle activity during extension of the spine from a flexed position, as compared to spinal flexion (Allen, 1948; Pauly, 1966; Ortengren and Andersson, 1977). Erector spinae muscle activity has also been shown to increase with corresponding increments in spine forward flexion until the "critical point" is reached at which they become inactive (Floyd and Silver 1951; Floyd and Silver, 1955). Since the lumbar spine flexed only slightly more during lifting than lowering (Figure 10), it is difficult to elucidate the contribution of this phenomenon from the *direction* difference illustrated in the erector spinae activity levels due to muscular contraction velocity.

Figure 26 illustrates that for the constrained condition, the load was handled at a greater peak horizontal distance from the ankle when lowering it as opposed to lifting it. This may be due to the force of gravity acting on the load once it starts to move during the lowering process. Subjects see that they must clear the wall and once they begin to move the load away from the body to do so, it's momentum carries it further away from the wall than needed. Thus, when handling a load with an obstruction in front of the body, the momentum of the load causes it's horizontal displacement from the body to overshoot when lowering it past the wall, as opposed to when lifting it.

Freestyle vs. Constrained Condition

Mean peak trunk flexion was observed to be 85.8° for the constrained condition and 65.7° for the freestyle condition. The increment in mean peak trunk flexion from freestyle to constrained conditions was solely due to an 18.4° increase in mean peak pelvic flexion, as the magnitude of lumbar spine flexion did not change (Figure 8). Ursulak and Potvin (1994) observed mean peak trunk flexion of 102.2° for stoop lifts. Mean peak lumbar spine and mean peak pelvic flexion increased from squat to stoop conditions by 13.1° and 31.9° respectively. Potvin *et al.* (1991) observed an average peak trunk flexion of 112° for stoop lifts. Average peak flexion of the lumbar spine and pelvis increased from squat to stoop conditions by 11° and 37° respectively. The diverse magnitudes of trunk flexion can be explained in terms of the techniques given for load handling. In the stoop condition, Ursulak and Potvin (1994) instructed subjects to try to perform the lifting tasks by flexing more with the trunk and less at the knees. Potvin *et al.* (1991) directed subjects to perform the stoop lifts with knees locked in a fully extended posture. In the current thesis, with a wall placed in front of each subject, and the freedom to perform the tasks in a manner most comfortable for them, subjects were able to flex the knees as well as the trunk, resulting in peak trunk flexion angles of a smaller degree than those for the instructed stoop techniques aforementioned. This was more consistent with actual workplace conditions. Integration of results from these three studies makes it apparent that required increments in trunk flexion from moderate magnitudes up to approximately 86° are accomplished primarily through greater flexion of the pelvis. Increments in peak trunk flexion to magnitudes greater than 86° are still attributed to that of the pelvis, yet the lumbar spine also further increases it's contribution. It must be noted that in cases where increments in peak trunk flexion are accomplished solely by the pelvis, sagittal plane motion between the lumbar spine and pelvis remains synchronous. The difference between the freestyle and constrained condition is that at a given angle of lumbar spine flexion there is greater pelvic flexion for the constrained condition (Figures 18 and 19).

The absence of any increase in peak lumbar spine flexion associated with increased peak trunk flexion plays a significant role in the differences in risks of low back injury between the two types of load handling conditions. It must be reiterated that progressive levels of lumbar spine flexion have been observed to induce increased strain on the interspinous and supraspinous ligaments (McGill, 1988). These ligaments have also been observed to be the first to sprain in cases of lumbar spine hyperflexion (Adams *et al.*, 1980). Hence, maintenance of the same lumbar spine posture from freestyle to constrained conditions did not impose any further strain on any of the posterior spinal ligaments aforementioned. Upon the demand of an increase in peak trunk flexion, it appears that individuals will do so by increasing peak pelvic flexion so as not to induce further strain on the posterior spinal ligaments. Thus, no increased risk of lumbar spine mechanical failure due to ligament sprain was associated with the increase in peak trunk flexion observed for the constrained condition.

Although the lumbar spine itself did not increase it's angle of flexion, the increase in peak trunk flexion did cause it's longitudinal axis to become more horizontal. This resulted in the surface of the lumbar intervertebral discs to become more vertical. Several studies have observed that this occurrence has caused a concomitant increase in the magnitude of the anterior shear force exerted on the lumbar intervertebral discs by the weight of the upper body and load (Park and Chaffin, 1974; Garg and Herrin, 1979; Potvin *et al.*, 1991). This would therefore increase the risk of low back injury associated with increased intervertebral shear forces.

Mean and peak TES and LES EMG levels were significantly greater for the constrained over the freestyle condition (Figure 9). This was due to the increase in mean peak trunk flexion observed in the constrained condition (Figure 8), causing an increase in the horizontal distance of the upper body and load from the spine's axis of flexion. Figures 24 and 25 illustrate the horizontal distance of the load from the ankle being greater for the constrained condition, for both the lift and the lower. This consequently resulted in a greater flexor moment about the thoracolumbar intervertebral joints for the constrained condition. Results indicate that the increase in flexion moment about the intervertebral joints was resisted by increments in back extensor muscle force generation. This has been mathematically illustrated to increase the compressive force on lower lumbar intervertebral discs (Anderson *et al.*, 1985; McGill and Norman, 1986). Hence, these intervertebral discs are at a greater risk of mechanical failure due to compressive forces associated with larger degrees of peak trunk flexion for the constrained condition.

The pelvic deviations observed for the freestyle condition are a phenomenon that have yet to be reported prior to this study. Figure 15 illustrates that to a certain extent, the sagittal plane angular displacement of the pelvis is dependant on the activity levels of both the thoracic and lumbar erector spinae musculature. Due to the direct attachment of the lumbar spine and the pelvis, pelvic rotation has a significant effect on the posture of the lumbar spine (Oliver and Middleditch, 1991). The following two phenomena are illustrated in Figure 15: (a) at the beginning of the freestyle lift, pelvic flexion caused the lumbar spine to extend approximately 2°, and (b) in the latter stages of the freestyle lower, pelvic extension caused an increase in lumbar spine flexion of approximately 8°. It must be noted that there was no significant *direction* effect for peak lumbar spine flexion within the freestyle condition, and therefore no difference in risk of low back injury.

The constrained condition showed no existence of pelvic deviations at the end range of motion (Figure 16). The reason behind this observation is that this condition required greater pelvic flexion than that for the freestyle condition (Figure 8). The pelvis did not flex at the beginning of the constrained lift as it had already accomplished that magnitude during trunk forward flexion to reach the load (Figure 17). The pelvis could not extend near the end of the constrained lower as further flexion was required in order to place the load back down to the floor (Figure 18). However, the following question is then raised: Why wouldn't the increase in erector spinae force levels at the beginning of the lift have caused a further increase in pelvic flexion, as was seen in the freestyle condition? There are two answers to this question. The first is that the greater mean peak pelvic flexion observed in the constrained lift may have caused it to reach near maximum flexion. Thus, the pelvis could not flex any further. The second is that the increase in mean peak pelvic flexion would have resulted in a greater flexor moment of the upper body and load about the hip joints. Hence, upon the initiation of the lift, greater activity levels of the pelvic extensors would have been required to meet this increase in moment demand. It is proposed that the activity levels of the pelvic extensors exceeded that of the erector spinae musculature by a substantial amount such that no further flexion could be induced on the pelvis. These explanations must be tempered with the fact that maximum pelvic flexion was not quantified and pelvic extensor EMG was not collected.

Chapter 6

CONCLUSIONS

The purpose of this study was to investigate the contributions of the lumbar spine and pelvis to several conditions of sagittal plane trunk motion while manually handling loads. Mean peak trunk, knee, lumbar spine, and pelvic flexion angles were measured for all conditions. Time-histories of lumbar spine and pelvic angular motion were also recorded throughout each trial. The thoracic and lumbar portions of the erector spinae musculature were monitored with surface electrodes to record their activity levels during the trials. Results of this study indicate a synchronous pattern of movement between the lumbar spine and pelvis during loaded sagittal plane trunk motion. However, the pattern of movement does become sequential when lifting a load up from the floor over a constraint. While the lumbar spine flexes to a greater magnitude during lifting, the pelvis does so during lowering of a load. Demands on the erector spinae musculature were shown to be greater during lifting as observed by their activity levels. Handling loads under conditions in which flexion is restricted at the knees causes an increase in peak trunk flexion, which is primarily an outcome of increased peak pelvic flexion. Erector spinae activation levels increase accordingly with greater angles of trunk flexion. No sex differences exist in lumbar spine - pelvic kinematics during manual materials handling. This is supported by the same trend observed in erector spinae muscular activation levels.

Hypotheses

Synchronous Movement of the Lumbar Spine and Pelvis

It was hypothesized that movement of the lumbar spine and pelvis would be synchronous during sagittal plane trunk motion while handling loads. This hypothesis was accepted for three of four conditions in this study. The pattern of angular motion appeared most synchronous for the constrained lower as pelvic deviations were observed during the freestyle condition. The pattern of motion between the lumbar spine and pelvis is sequential when having to lift a load from the floor over a constraint.

Sex Effects

It was hypothesized that males would accomplish the tasks with greater peak flexion of the lumbar spine, while females would do so with greater peak pelvic flexion. This hypothesis was not accepted. No sex differences were detected for either of these two variables. Data were supported by the similar trend seen in mean and peak TES and LES EMG levels.

Lift vs. Lower Effects

Peak lumbar spine flexion was hypothesized to be greater at the end of the lower than at the start of the lift. Results indicate the opposite: peak lumbar spine flexion was greater at the start of the lift than at the end of the lower. Individuals therefore appear to be at a slightly higher risk for sustaining a low back injury during lifting than lowering of loads.

Freestyle vs. Constrained Lifting Effects

(a) Peak Trunk Flexion

With the wall placed in front of the subjects, it would restrict flexion at the knees so it was hypothesized that there would be an increase in peak trunk flexion to compensate. This hypothesis was accepted as peak trunk flexion was significantly greater for the constrained condition versus the freestyle condition. This was supported by significantly greater mean and peak levels of TES and LES EMG observed for the constrained condition.

(b) Peak Lumbar Spine Flexion

It was hypothesized that the increase in peak trunk flexion for the constrained condition would be partly due to an increase in peak lumbar spine flexion. Furthermore, it was proposed that constrained peak lumbar spine flexion would reach a certain magnitude that would elicit the flexion-relaxation phenomenon (FRP). This would have been illustrated by the silencing of the LES EMG signal. Results indicate that peak lumbar spine flexion did not increase from freestyle to constrained conditions. The LES EMG signal did not show any period of silence, therefore signifying the absence of the FRP.

(c) Peak Pelvic Flexion

It was hypothesized that although peak lumbar spine flexion would increase somewhat, the main contribution to the increase in peak trunk flexion for the constrained condition would come from the pelvis. This hypothesis was accepted as the significant increase observed in mean peak pelvic flexion was the sole contributor to the augmentation in mean peak trunk flexion for the constrained condition. It must be noted, however, that in cases where increments in peak trunk flexion are accomplished solely by the pelvis, sagittal plane motion between the lumbar spine and pelvis remains synchronous.

Recommendations for Future Research

One consideration to take into account is that there may exist a load effect on the pattern of movement between the lumbar spine and the pelvis. It is possible that loads of greater magnitude may create flexor moments about the lumbar intervertebral joints that may exceed the capabilities of the erector spinae musculature at extreme angles of flexion. With this, it may be possible that extension of the pelvis must initiate trunk extension during lifting until the flexor moment can be balanced by the back extensors. Hence, a study could be performed at varying load magnitudes in the attempt to determine if these variations alter the pattern of sagittal plane displacement between the lumbar spine and pelvis.

Collecting myoelectric signals from other muscles may also help to further explain some of the phenomena observed in the current study. Studying the activity levels of the gluteus and hamstring muscles may help to explain some of the observed effects on mean peak pelvic flexion. However, one must be cautioned in the interpretation of this data due to the biarticular nature of these muscles. The use of indwelling electrodes may have also provided information on activity levels of deeper muscles such as the psoas. This muscle has been observed as one that stabilizes the lumbar spine during various tasks (Crisco and Panjabi, 1990; Santaguida and McGill, 1995). Although no sex effect existed for peak lumbar spine flexion or the associated EMG variables, sex differences in mean psoas EMG levels may reveal a difference in the requirement for spinal stability at a given level of flexion.

Another route for future research could be whether the element of fatigue affects the pattern of movement between the lumbar spine and pelvis while handling loads. A protocol could be developed in order to fatigue the muscles associated with trunk flexion and extension during manual materials handling, and to determine whether their localized magnitudes of fatigue alter the synchronous pattern of movement between the lumbar spine and pelvis. One could also investigate into whether this would also affect the occurrence of the pelvic deviations observed in the freestyle condition. This type of study would have significant implications towards injury risk differences at the start and end of a work-shift.

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APPENDIX A

Information and Consent Form

Study Title: Relative contributions of the lumbar spine and pelvis to trunk motion during sagittal plane manual materials handling tasks.
Conducted by: Christopher McKean (M.Sc. student) and Dr. James Potvin (Department of Human Biology and Nutritional Sciences)

Study Details

The purpose of this study is to monitor the contributions of the lumbar spine and pelvis to trunk sagittal plane motion while lifting and lowering loads. The study will require each subject to attend one main session approximately 1 hour in duration. Participants will perform 10 lifts and 10 lowers of a specific load (males = 15 kg, females = 8.5 kg) for two conditions: freestyle and constrained. The constrained condition simulates an industrial bin in which the worker must reach over the side to handle the load. Subjects are instructed to lift/lower the load in a manner most comfortable to them. Reflective markers will be placed on the skin at 10 sites: ankle, knee, greater trochanter of the femur (hip), iliac crest, shoulder, elbow, wrist, T1 spinous process, L1 spinous process, and the sacrum. Movement of these markers will be tracked throughout the trials via videography. Surface electromyography will be collected unilaterally from 2 sites: lumbar erector spinae, and thoracic erector spinae muscles. A slight skin irritation may arise from the reflective markers and the electrodes. Maximum voluntary contraction of the two muscles being studied will be required via forceful trunk extension. Muscle stiffness may result following the collection session, but should be no more than what might be experienced after any unaccustomed physical activity.

Consent of Subject

I have read and understand the information presented above for the procedures and risks involved in this study, and have received satisfactory answers to questions related to this study. The specific details have been explained. I understand my identity will be protected throughout my participation in this study. I am aware that I may report what I consider violations of my welfare to the Office of Research, University of Guelph, and withdraw from the study at any time. With full knowledge of the foregoing, I agree of my own free will to participate as a subject in this study.

Date: ______________________________

Participant's name (print): ______________ Participant's Signature: ______________

Experimenter: ______________ Experimenter's Signature: ______________

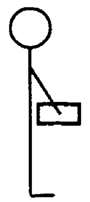
APPENDIX B

Fin Angle Calculations

Below is a detailed description of the methodology used to calculate the angle obtained by any of the triangular, 3 markered fins.

Posture:

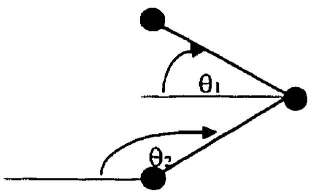
(a) Erect Stance



(b) Full Forward Flexion

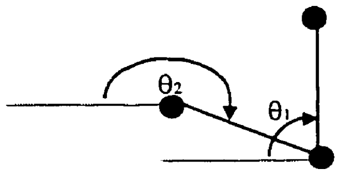
FIN ORIENTATION ON DORSUM OF SUBJECT:

(a)



$\theta_1 = 35°$
$\theta_2 = 135°$
$\theta_F = (35° + 135°)/2$
$= 85°$

(b)



$\theta_1 = 90°$
$\theta_2 = 200°$
$\theta_F = (90° + 200°)/2$
$= 145°$

The nature of the angles subtended between two markers on the triangular fins causes both θ_1 and θ_2 to increase with trunk forward flexion. As θ_F is calculated as a running average of θ_1 and θ_2, consequently, it too increases with trunk flexion. The methodology of these calculations is consistant for all triangular fins, therefore causing all fin angles to increase with trunk forward flexion.

APPENDIX C:

PRESENTATION OF ANOVA INTERACTIONS

Interactions

No two way interactions involving *sex* were significant.

Condition x Direction

The *lifting condition* x *direction* interaction was significant for peak trunk flexion ($p < 0.001$), peak knee flexion ($p < 0.05$), mean thoracic EMG ($p < 0.05$), and Peak Thoracic EMG ($p < 0.01$). The 4 pairwise *lifting condition* x *direction* means comparisons are presented for all of these variables in Tables 4(a) and 4(b).

Table 4. Pairwise *condition* x *direction* contrasts by means comparisons for **(a)** Peak Trunk Flexion Angles and Peak Knee Flexion Angles, and **(b)** Mean Thoracic and Peak Thoracic EMG levels. Contrasts were A vs. B, with A or B in the Larger (Lrgr) column to indicate which cell was greater in magnitude (* $p < 0.001$ unless stated otherwise). Differences in A-B are expressed in degrees for peak flexion angles and in %MVC for EMG variables.

(a)

Contrast		Peak Trunk Flexion			Peak Knee Flexion		
A	**B**	**Lrgr**	**A - B**	**p <**	**Lrgr**	**A - B**	**p <**
LiftC	LiftF	A	17.41	*	B	-38.12	*
LowC	LowF	A	22.70	*	B	-41.12	*
LiftC	LowC	B	-2.92	0.01	A	5.18	*
LiftF	LowF	A	2.37	0.05	A	2.18	0.05

(b)

Contrast		Mean Thoracic EMG			Peak Thoracic EMG		
A	**B**	**Lrgr**	**A - B**	**p <**	**Lrgr**	**A - B**	**p <**
LiftC	LiftF	A	5.27	*	A	12.88	*
LowC	LowF	A	3.21	*	A	6.87	*
LiftC	LowC	A	9.81	*	A	18.45	*
LiftF	LowF	A	7.75	*	A	12.44	*

APPENDIX D:

KINEMATIC AND PHASE ANGLE VARIABILITY

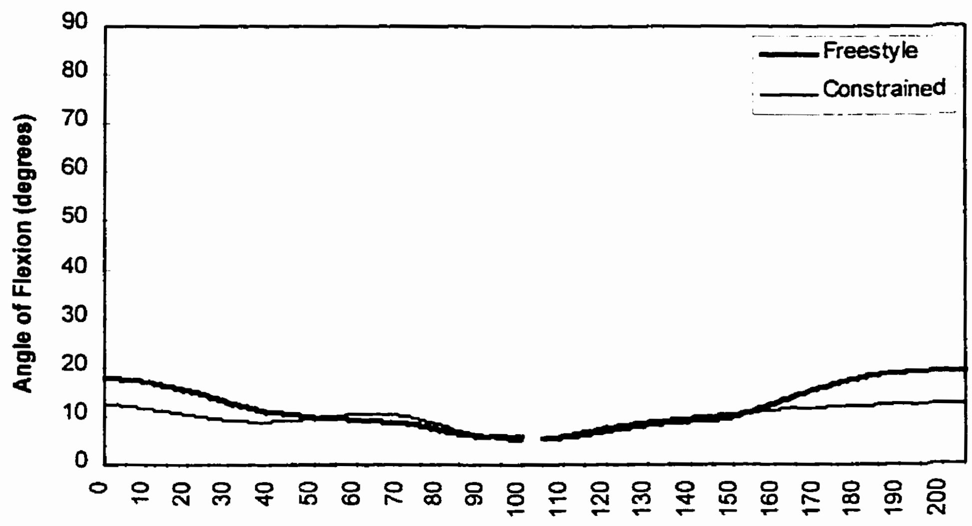


Figure 27. Illustration of Trunk flexion standard deviation throughout all four *lifting condition* x *direction* combinations. Data are poooled across *sex* (n = 22).

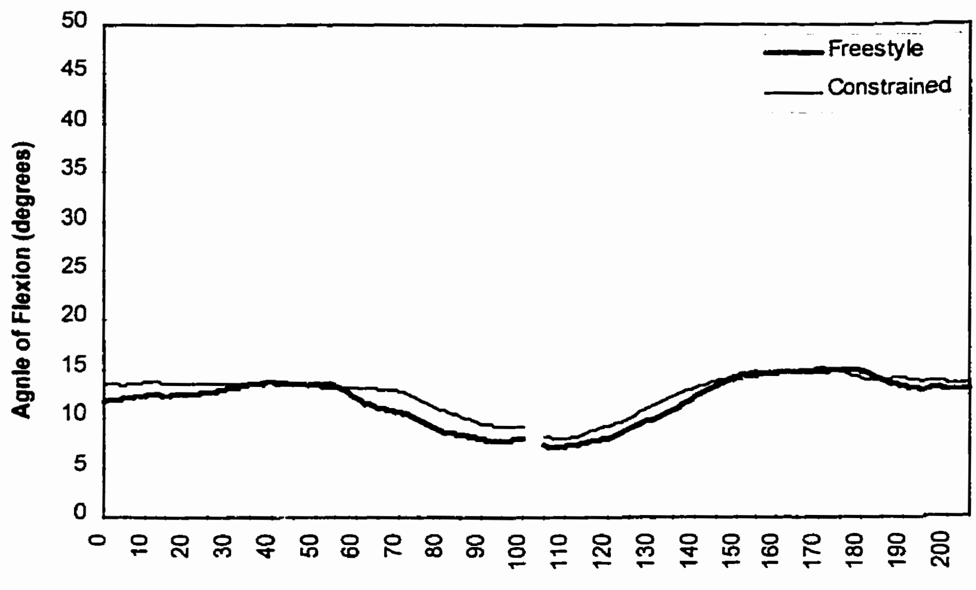


Figure 28. Illustration of Lumbar Spine flexion standard deviation thoughout all four *lifting condition* x *direction* combinations. Data are pooled across *sex* (n = 22).

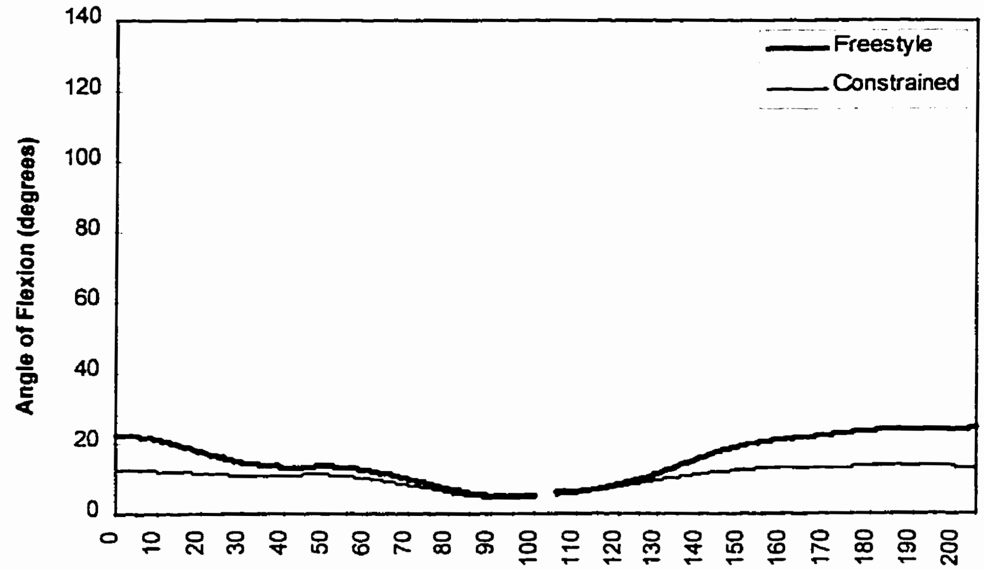


Figure 29. Illustration of Knee flexion standard deviation thoughout all four *lifting condition* x *direction* combinations. Data are pooled across *sex* (n = 22).

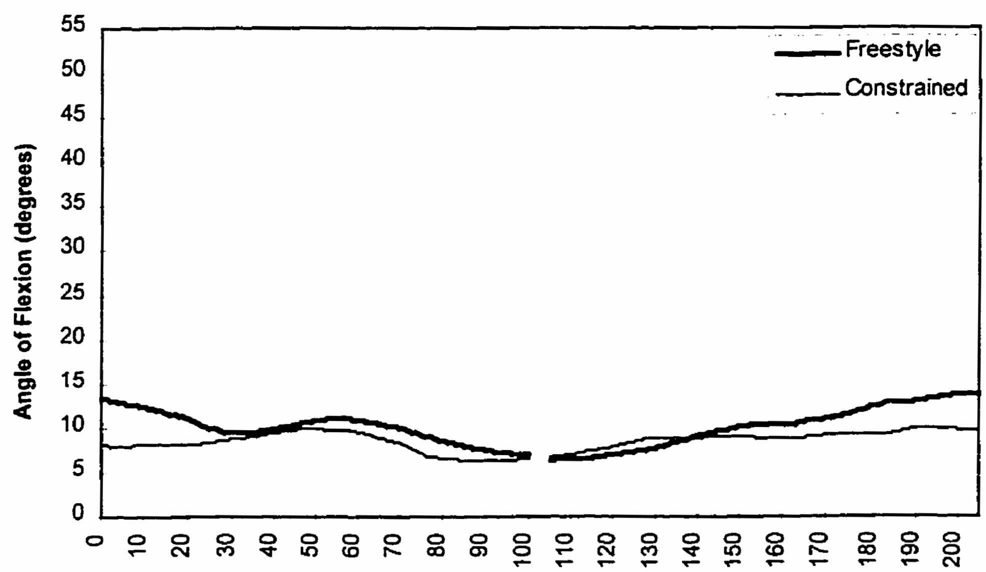


Figure 30. Illustration of Pelvis flexion standard deviation thoughout all four *lifting condition* x *direction* combinations. Data are pooled across *sex* (n = 22).

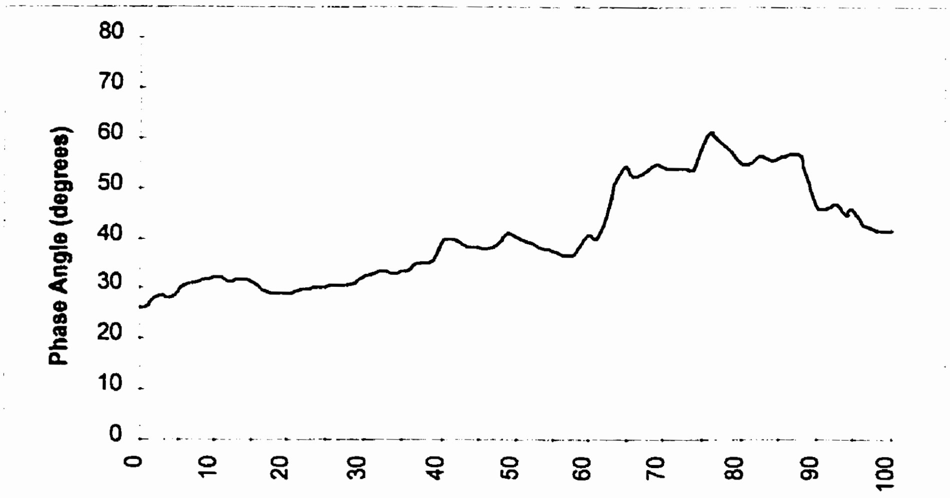


Figure 31. Illustration of Phase Angle standard deviation throughout the Freestyle Lift. Data are pooled across *sex* (n = 22).

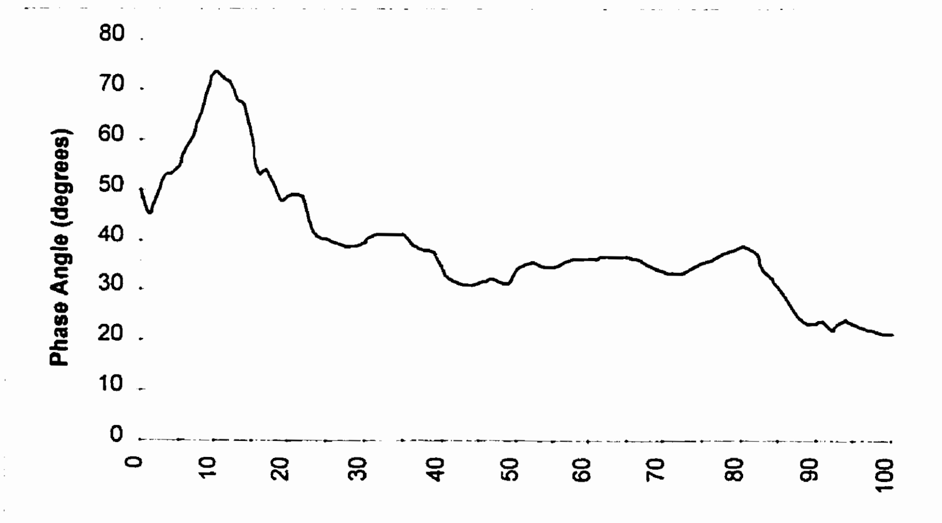


Figure 32. Illustration of Phase Angle standard deviation throughout the Freestyle Lower. Data are pooled across *sex* (n = 22).

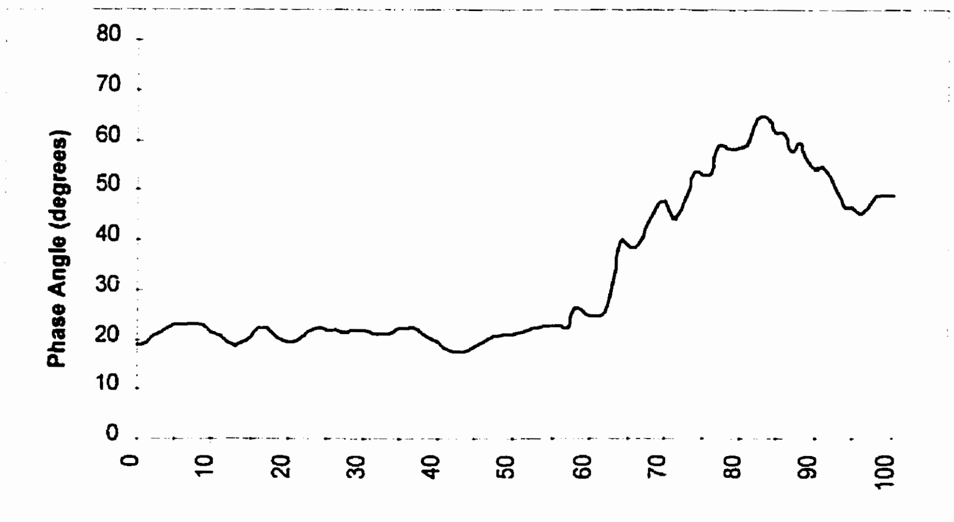


Figure 33. Illustration of Phase Angle standard deviation throughout the Constrained Lift. Data are pooled across *sex* (n = 22).

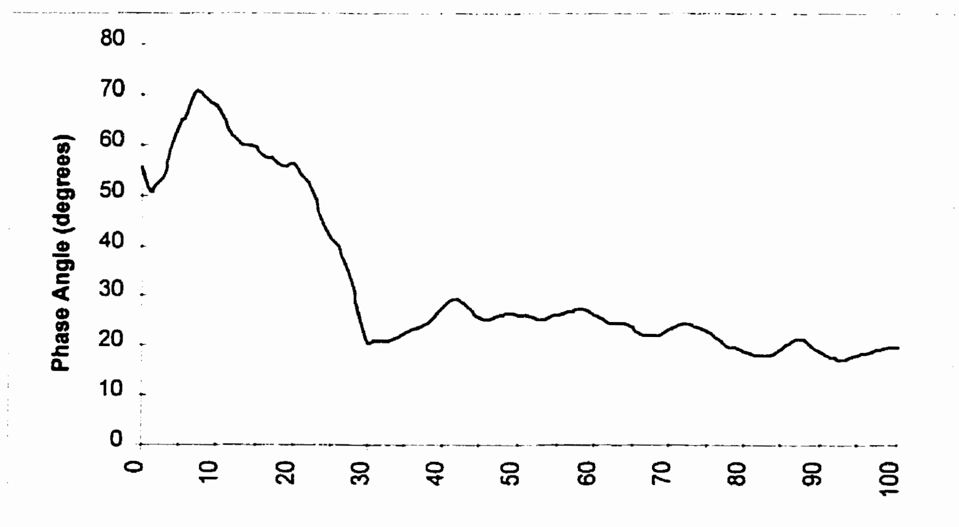


Figure 34. Illustration of Phase Angle standard deviation throughout the Constrained Lift. Data are pooled across *sex* (n = 22).

APPENDIX E:

GRAPHIC ILLUSTRATION OF THE CALCULATION OF PHASE ANGLE BETWEEN THE LUMBAR SPINE AND PELVIS

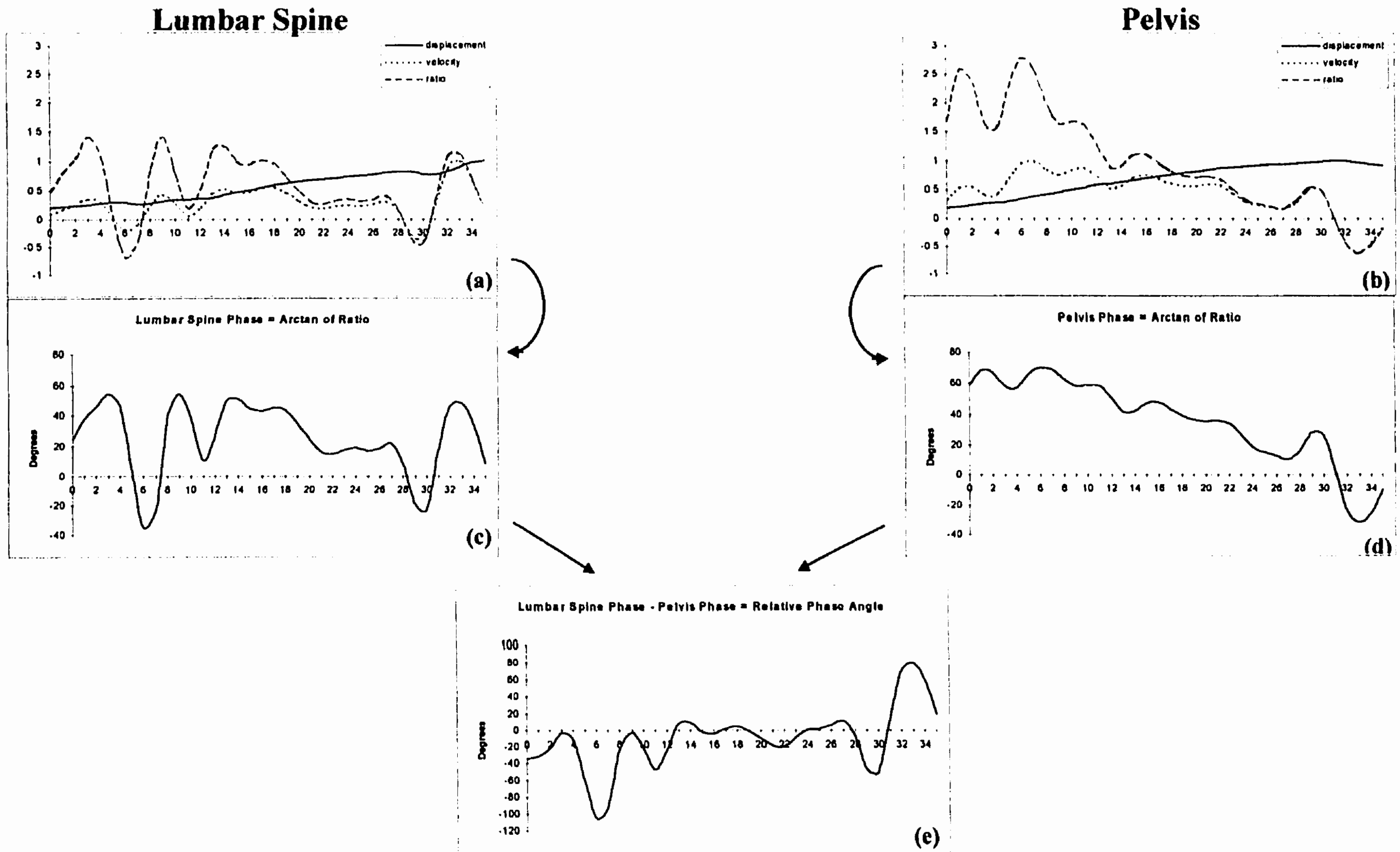


Figure 35. Graphic representation of calculations incorporated into the Relative Phase Angle between the Lumbar Spine and Pelvis, for a typical trial of a constrained lower. Normalized angular displacements, velocities, and ratios of normalized velocities to normalized displacements are in (a) and (b). The phase of each segment is calculated as the arctangent of the calculated ratio, and are presented in (c) and (d). The Relative Phase Angle is calculated as the difference between the Lumbar Spine Phase and Pelvis Phase, and is illustrated in (e).