The Effect of a Commercially Available Abdominal Support Belt on Torso Posture, Lift Strength, and Spinal Compression

by

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ABSTRACT

Intervention programs have been developed to reduce the occurrence and recurrence of back pain and to prevent low-back injury. One program involves the wearing of abdominal support belts when manual materials handling activities are performed. However, poor postures and restricted ranges-of-motion may result due to belt usage. Unfortunately to date, the majority of research available regarding support belt effects only address those supports designed for therapeutic reasons. The primary purpose of this research was to measure changes in posture when a commercially available support belt was worn. The support belt used in this investigated was the Decade Back Support manufactured by Chase Ergonomics, Incorporated. In addition, this study investigated whether the Decade belt affected static lift strength and predicted spinal compression of the L3/L4 intervertebral disc.

Eight males and eight females, aged 23.6 ± 2.6 and 22.4 ± 4.3 years respectively, were asked to perform maximal static exertions on handles attached to a steel rig. The exertions differed in symmetry and handle height while the support belt was worn and not worn. Posture data was collected through a WATSMART 3-D motion analysis system. The vertebra prominens body landmark was examined to determine if there was a change

in flexion and lateral bending angle of the torso across Belt conditions. Axial twist angle between the shoulders and the hips was also investigated. Strength data was collected through a force platform across three orthogonal axes. Spinal compression was predicted through the use of the Minimum Intensity Compression (MIC) biomechanical model proposed by Bean, Chaffin, and Schultz (1988).

It was found that torso posture, with the exception of axial twist, and static strength were not significantly affected by belt use. Axial twist for low asymmetric exertions was significantly larger when the belt was worn as compared to when the belt was not worn. This result may have been caused by the belt not allowing the hips to turn with the upper torso when the low exertions were performed. Maximal strength exertion was not affected by belt use. Predicted spinal compression was significantly lower when the belt was worn (2737.87 N) as compared to the nonbelt conditions (3087.47 N).

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1. INTRODUCTION

1.1. Rationale

Manual materials handling, specifically those tasks involving lifting, has been recognized as a major hazard to industrial workers (Andersson, Pope, and Frymoyer, 1984). Up to 35% of sedentary workers and 47% of physical laborers may acquire occupationally related low-back pain (Rowe, 1969). Approximately 2% of the total United States industrial work force suffers a compensable back injury each year (Beals, 1984). Once an individual has experienced an episode of back pain or impairment, the risk of subsequent injury is increased. It is also likely that each recurrence will be more severe. Walsh and Schwartz (1990) report that estimates for the lifetime incidence of low-back pain in Americans range from 65 to 85%. The number of work days lost because of back pain is approximately 1.4 days per worker per year (Andersson, 1991). These lost workdays constitute nearly one-quarter of all disabling work-related injuries in the United States.

Intervention programs have been developed to reduce the occurrence and recurrence of back pain and to prevent low-back injury. One such program is the use of abdominal support belts during manual materials handling activities. According to a survey administered by the Center for Workplace Health (1993) of over 400 commercial enterprises throughout North America, 77% indicated they have incorporated the used back supports as part of their safety program. When broken down by industry, 88% of the respondents who used back supports were classified in the industrial, manufacturing, and service sectors. It is in these sectors where manual materials handling can be a major component in operations.

Past research has dismissed (Hemborg and Moritz, (1985), McGill and Norman, (1987), McGill, Norman, and Sharratt, (1990)) or advocated (Bartelink, (1957), Morris, Lucas, and Bresler, (1961), Grew and Deane, (1982), Gracovetsky, Farfan, and Helleur (1985), Harman, Rosenstein, Frykman, and Nigro, (1988), Lander, Simonton, and Giacobbe, (1989)) conclusions regarding the effectiveness of abdominal support belts for generating increased intra-abdominal pressure (IAP). Of possibly greater importance are support belt posture effects. Some researchers have recently proposed that use of abdominal support belts may impose an increased risk of injury as a result of influencing posture and range-of-motion (Harman, Rosenstein, Frykman, and Nigro (1988), Lander, Simonton, and Giacobbe (1989), and Reddell (1990)). However the majority of available research that addresses posture effects has only tested supports that were designed for therapeutic purposes. This available research has provided little guidance to industrial safety personnel, employers, and employees regarding the effectiveness of commercial back supports as personal protective equipment. As a result, the need for research regarding the physical effects of commercially available support belts is great.

1.2. Objectives

Postural effects due to the use of a commercially available abdominal support belt during various static lift exertions were investigated. The primary experimental objectives of this study were as follows:

Objective 1: To develop a procedure and measure the difference in torso posture across static lift exertions differing in symmetry and height across the belt and nonbelt conditions of a commercially available abdominal support belt.

Objective 2: To measure and compare the static lift strength across the belt and nonbelt conditions.

Objective 3: To predict and compare the levels of low-back compressive force at the L3/L4 vertebral disc across both belt and nonbelt use conditions using a three-dimensional biomechanical model.

2. LITERATURE REVIEW

2.1. Overview

This chapter summarizes the literature available regarding abdominal support belts and similar devices and their influence on the body. It is presented in seven sections. Section 2.2. presents a review of the muscular anatomy of the lower back and torso. Section 2.3. addresses the history of abdominal support belts and devices and their hypothesized effect upon the body. Section 2.4 presents a chronological review of the available literature regarding the investigation into the general effects of the use of abdominal support belts on the body. Section 2.5. narrows the literature discussion to those studies focusing upon the effects of the use of abdominal support belts on lift posture. Section 2.6. discusses IAP and the pros and cons of inclusion in current biomechanical models. Section 2.7. discusses biomechanical models and the model used in this research. Section 2.8. discusses the differences in commercially available abdominal support belt designs. A summary of the literature is then presented.

2.2. Review of the Muscular Anatomy of the Lower Back and Torso

Twenty-two intervertebral disc joints and 24 paired facet joints make up the human spine. They are grouped in the cervical, thoracic, lumbar, and sacrum regions which are based upon the curvature of the spine when not in flexion or extension. The cervical region is made up of 7 vertebrae, the thoracic region is made up of 12 vertebrae, and the lumbar region is made up of the remaining 5 vertebrae. The sacrum consists of five fused vertebrae (the intervertebral discs in the sacrum region are ossified). Both the cervical and lumbar regions are in lordosis curvature while the thoracic and sacrum regions are in kyphosis curvature. Spinal curvature can be affected by posture, physical activity, obesity, pregnancy, trauma (cumulative or acute), and/or disease.

There are five main muscle pairs which produce extension, flexion, and twisting of the spine. They are the (1) rectus abdominis, (2) internal obliques, (3) external obliques, (4) erector spinae, and (5) latissimus dorsi. Each muscle pair is made up of a left and right component (Figures 1 through 3). Of the five different muscle pairs, the rectus abdominis and the internal and external obliques are located anterior to the spine while the erector spinae and latissimus dorsi pairs are located posterior to the spine. The rectus abdominis muscles rise from the pubic crest and join with the lower costal cartilages and xiphoid process (sternum). The internal and external obliques roll away from the midline of the torso with the external obliques located on top of the internal obliques. All three muscle pairs act to compress the abdominal region during expiration, urination, and defecation. They also assist in maintaining pressure on the curvature of the low-back, resisting excess lumbar lordosis and extension of the lumbar low-back area.

The main extensors of the spine are both the erector spinae and the latissimus dorsi muscle pairs. The erector spinae pair is oriented vertically along the longitudinal axis of the back. In the lumbar region, they are thick and uniform. Above the lumbar region, the erector spinae muscles split into three smaller and thinner bundles attaching to the ribs (iliocostalis), upper

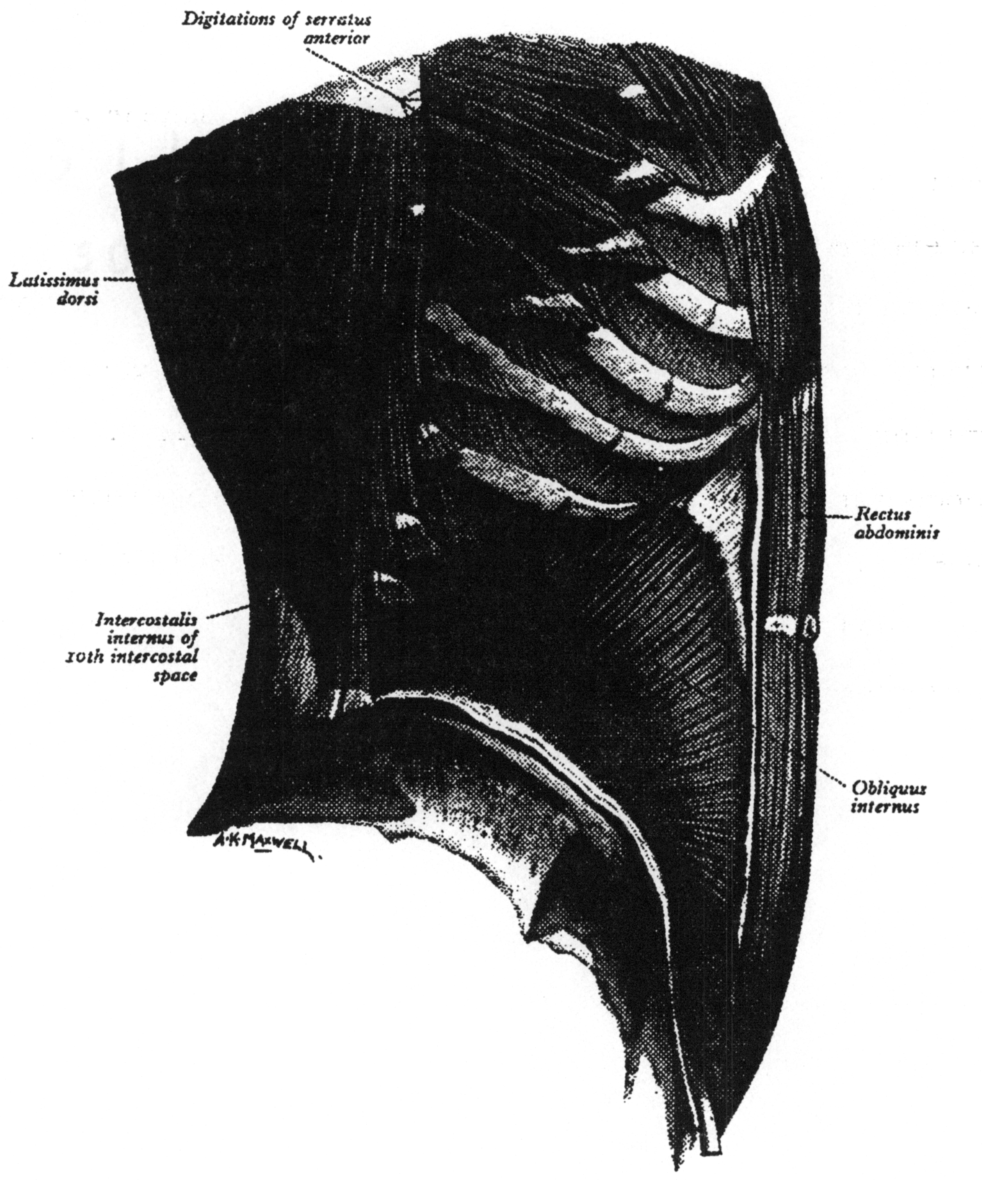

Figure 1. Right sagittal view-trunk muscle system. Taken from Hollinshead (1976).

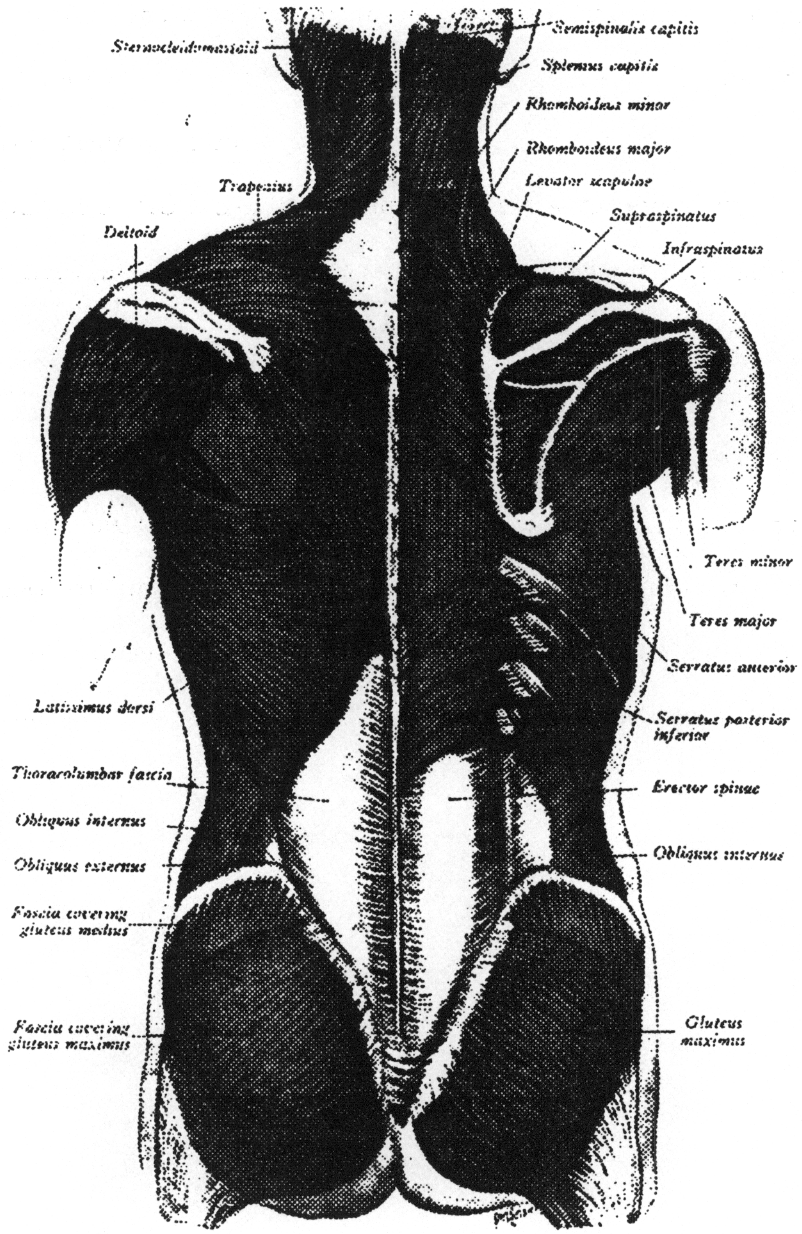

Figure 2. Posterior view-trunk muscle system. Taken from Hollinshead (1976).

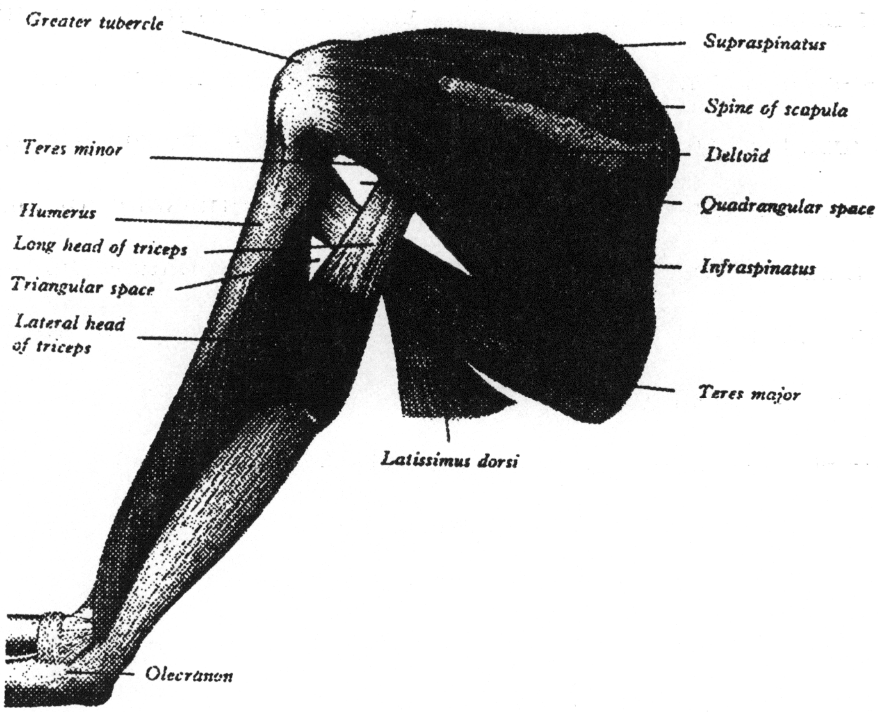

Figure 3. Muscular system of the shoulder. Taken from Hollinshead (1976).

vertebrae (longissimus), and the head (spinalis). Each latissimus dorsi muscle is located anterior to the humerus (behind each shoulder) and thus has an excellent mechanical advantage for medial rotation of the shoulder and extension of the spine.

2.3. Intra-Abdominal Pressure and Support Belts

The earliest known back support belts and lumbosacral corsets date back to the Minoan period approximately 2000 B.C. in which they were used to enhance the female figure. During the time period of approximately 900 A.D., pre-Columbian Indians used tree bark corsets as a method of reducing back pain. In the 12th century surgeons developed cumbersome spinal braces made from metal and wood. The first uses of external abdominal supports, such as braces, plaster-of-Paris jackets, corsets, and belts, in the United States provided treatment of low-back injuries and disabilities such as Pott's disease, scoliosis, and spinal tuberculosis (Evans, 1921). In 1923 it was first proposed by Sir Arthur Keith that pressure within the trunk cavity might provide assistance in lifting and increased mechanical efficiency.

As a load is lifted by the body in the frontal plane, a flexion moment is generated about the lumbar spine. As a reflex to this moment, the abdominal and thoracic cavities pressurize by tensing the diaphragm and abdominal wall muscles. This pressurization acts upon the front of the lumbar spine producing an extensor moment. It is hypothesized that there is a reduction in spinal stress and compression because the extensor muscle forces required for equilibrium are reduced due to the generation of this intra-abdominal pressure (IAP) extensor moment (Chaffin and Andersson, 1991).

To aide the diaphragm and abdominal wall muscles in the generation of IAP, it is hypothesized that an external belt may be used. When worn, the abdominal support belt helps stiffen the trunk region of the body. Use of these belts are believed to result in three main effects in relieving lumbar spinal compression. It is believed that support belt usage will result in (1) a shifting of a portion of the load of the upper part of the body from the spine to the rest of the trunk by producing abdominal compression, (2) providing sufficient support to allow relaxation of the muscles of the trunk, and (3) providing a placebo effect in reminding wearers of proper lifting techniques and postures. Support belts may also reduce the forces on the trunk directly by supporting some of the load due to the physical construction of the belt. In addition, abdominal belts have been suspected of acting as a splint, reducing the range-of-motion, and thereby affecting posture. A variety of different abdominal support belts of varying construction are commercially available on the present day market.

2.4. Previous Research Regarding the General Effects Resulting from the Use of Abdominal Support Belts

2.4.1. Waters and Morris (1970)

This study was designed to compare the electrical activity of certain trunk muscles during standing and walking, with and without spinal supports. Both a lumbosacral corset and a chairback brace were investigated using male and female participants. The experimental sessions consisted of collecting EMG recordings from the right side of the body. The specific muscles studied were the longissimus dorsi, iliocostalis lumborus, iliocostalis dorsi, multifidus, rotatores, rectus abdominis, oblique externus abdominis, and the oblique internus abdominis. In addition, heel strike data was collected by means of electrical switches attached to the participants' shoes.

Each participant was fitted with each of the supports and the electrodes were placed in the proper position for each of the muscles being observed. The participants first stood at rest, then walked on a level treadmill at 4.39 kilometers per hour, at 5.29 kilometers per hour, and finally at 4.39 kilometers per hour up a 5-degree incline. Test sequences were made with the participants standing and walking with and without the spinal supports.

Data interpretation revealed that when the participants were at rest, both supports either decreased or had no effect on the electrical activity of the back muscles for a majority of the subjects. At the slower speed, neither support had any significant effect on the activity observed. At the higher walking speed, the activity of the back muscles was increased for a majority of the subjects in comparison with the activity of those muscles when the chairback brace was worn. Waters and Morris (1970) suspected the effect on electrical activity was due to the more rigid physical construction of the belt. They believed the greater electrical activity observed was due to the muscular exertion of the back muscles as they attempted to overcome the immobilizing effect of the chairback brace. In addition, it was noted through the interpretation of heel strike data, each support did not interfere with the normal function of the lower extremities during walking.

2.4.2. Kumar and Godfrey (1986)

This investigation focused on performing a comparative evaluation of six commonly prescribed spinal supports. The supports studied were the sacroiliac belt, the lumbosacral corset, Harris brace, the Macnab brace, the Knight brace, and the Taylor brace. Twenty male and female participants were asked to perform sagittal, lateral, oblique lifts, and same level side-to-side weight transfers. The dependent variable in this study was measured peak and sustained IAP.

Each of the lifts consisted of movement of the weight from ground to knee, ground to hip, and ground to shoulder. The loads weighed 7 and 9 kg. IAP was measured through the use of a pressure sensitive radio pill.

The results obtained by Kumar and Godfrey (1986) indicate no significant difference in IAP changes due to the use of the different spinal supports in both males and females. They did, however, note that supported conditions were statistically different from one another for every lifting activity. From these results, the authors concluded there is no difference in the supports studied but they subjectively observed that the different braces restricted the torso movement differently. It was also noted that the magnitude of the IAP for sagittally symmetrical activities was significantly lower than the lateral and oblique lifts.

2.4.3. Harmon, Rosenstein, Frykman, and Nigro (1989)

This study involved the measurement of both ground reaction forces (GRF) and IAP during the use of abdominal supports belts while participants repeatedly lifted barbells at 90% of their maximum voluntary lift. Each of the subjects performed the dead lift for this experiment and had been coached to ensure that they would lift according to a standard technique. IAP was measured by means of a catheter pressure transducer inserted through the nostril.

Results indicated that IAP and GRF rose sharply once the lift was initiated after which GRF plateaued and IAP either plateaued or declined. With regards to belt use, IAP rose more quickly when a belt was worn compared to the nonbelt condition. It was also observed that IAP peak values were significantly greater with a belt than without a belt. In contrast, average IAP values for the entire lift were significantly lower compared to the nonbelt condition.

From these results Harman et al. (1989) concluded that a support belt increases IAP during the dead lift exercise. They also hypothesized the belt may prevent protrusion of the abdomen by forcing the abdominal muscles to move inwardly counteracting more directly the weight of lift.

2.4.4. Hunter, McGuirk, Mitrano, Pearman, Thomas, and Arrington (1989)

Hunter et al. (1989) were interested in the effect of wearing a 10-cm weight belt on blood pressure and heart rate while performing dead lifts, bicycle riding, and one-armed bench presses. Five males and one female participated in the experiment.

For the dead lift, a load of 40% of each participant's maximum weight was held in a lift posture for two minutes. The participants were instructed to breathe throughout the trials so breath holding effects were not confounded. Blood pressure and heart rate were significantly higher during the lifting conditions when the belt was worn. Hunter et al. (1989) concluded that individuals who have potential cardiovascular system problems are probably at greater risk when undertaking exercise while wearing a back support.

2.4.5. Walsh and Schwartz (1990)

This study was designed to investigate the effect of multimodal intervention in the prevention of back injury and to evaluate the potential adverse effects of using a lumbosacral corset in the workplace. Intervention techniques included abdominal belt use, educational "back care" instruction, or a combination of both. Ninety warehouse workers (from a grocery chain) participated in the study.

Each of the workers was randomly assigned to one of three treatment groups: group 1 had no intervention; group 2 received only a training session in back pain prevention and body mechanics on the job; and group 3 received the training session and were asked to wear an abdominal support belt only during working hours for a period of 6 months. In addition to a questionnaire solicited by the experimenters at the beginning and end of the experiment, abdominal strength was assessed through isometric contraction measurement. Work injury incidence data was also collected.

Results indicated that of the 90 workers who participated, follow-up information was obtained from 91% of those workers. Those who were assigned to group 3 missed a significantly less number of work days during the 6 month period as compared to the previous 6 months before the study. Questionnaire knowledge for both group 2 and group 3 was significantly improved compared to the control group. With regards to the abdominal strength measurements, no significant differences were observed when all three groups were compared. In addition, productivity gains and the injury rate showed no significant differences between groups.

As a result of this investigation, Walsh and Schwartz (1990) concluded that the wearing of a back support reminds the user of proper body mechanics and lifting techniques. They also concluded that atrophy of the muscles that support the spine would not occur as a result of prolonged use. In addition, they support the position that the use of appropriate education and physical support will result in decreased lost time from work due to back injury.

2.4.6. McGill, Norman, and Sharratt (1990)

The purpose of this study was to determine if weightlifting belts have an effect on trunk muscle activity and IAP. Six participants lifted loads (between 72.7 and 90.9 kg) both with and without wearing a weightlifter belt. In addition, participants lifted both with their breath held or continuously expiring during the lift. McGill et al. (1990) were interested in whether a belt could assist in elevating IAP in the absence of elevated abdominal muscle activity. They were also interested in whether the demands on the low-back musculature were reduced as a result of the trial conditions.

The lifting apparatus only allowed the load to travel in a vertical track and feet placement was controlled. The belt was a standard four inch leather support belt. The experiment was repeated for two participants who wore a commercially available support belt. In addition a national class weightlifter performed three dead lifts of loads of 187, 230, and 243 kg. IAP measurements were made with a pressure catheter placed in the stomach. Dynamic vertical hand forces were recorded by strain gauged handlebars on the apparatus.

Activity of the rectus abdominis and erector spinae muscles was measured through the use of EMG electrodes.

With regards to belt effects, it was observed that the competition support belt slightly increased IAP but no appreciable differences were observed in rectus abdominis or erector spinae activity. McGill et al. (1989) concluded that belts do not appear to contribute to support of the loaded lumbar spine by means of increased IAP. McGill et al. (1989) also reported that participants expressed a feeling of security and stability from wearing a belt even though the data suggested that spinal compression was not reduced. They hypothesized that the belt provided a means of resistance to bending of the trunk by its physical construction. In addition they believed the belt reminded the lifter to avoid lumbar flexion and to follow proper lifting techniques.

Among the other effects investigated, breath holding appears to unload the lumbar spine and this effect is not significantly altered by the wearing of a support belt.

2.4.7. Lander, Hundley, and Simonton (1992)

This study examined the effectiveness of a weightlifting belt in performing multiple repetitions of the squat exercise. GRF, IAP, and EMG activity of the external obliques, erector spinae, vastus lateralis, and bicep femoris muscles were recorded from five different skilled participants who were all engaged in weight training programs. The participants were asked to perform eight consecutive lifts at 75-80% of the weight of their one repetition maximum effort. In addition, cinematographic data was recorded to determine if lifting technique varied across the trial conditions.

It was observed that the lifters showed no appreciable differences in technique across repetitions excluding the time it took to complete the eight lifts. Lifting while wearing the belt was generally performed faster than lifting without a belt. In addition, IAP values were approximately 25-40% higher when the belt was worn compared to when it was not worn. EMG data indicated that no differences were observed for the erector spinae and external obliques. However differences in muscle activity were observed for the vastus lateralis and bicep femoris muscles with the belt use conditions being significantly greater. These results suggest the participants used greater knee and hip extensor musculature with a weightlifting belt than without a belt.

The authors concluded the use of a weightlifting belt appears to provide an added degree of protection during submaximal repetitious lifting. The authors also suggest that loading on the spine is reduced due to the use of a belt. This aspect, however, was not investigated.

2.5. Previous Research Regarding Abdominal Support Belts and Lifting Posture

2.5.1. Norton and Brown (1957)

Different support devices which were usually prescribed in the treatment of low-back disabilities were investigated with regard to their effect upon posture and motion of the lumbosacral spine. The different braces tested included a plaster-of-Paris jacket, the Goldthwait, the Taylor (both rigid and flexible construction), the Arnold-Abbott, the Jewett, the chairback, an experimental design developed by the research team, and the Williams brace. Males participated in this research.

Lateral view roentgenograms were taken while the participants were standing and bending during while wearing each brace. Conditions varied by the amount of forward flexion. During each condition, angles were measured by the intersection of lines drawn on the roentgenograms through the tips of the two most anterior projections of the anterior surface of each adjacent lumbar vertebra. The sacral region of the spine was handled in the same manner.

The second method involved the use of Kirschner-wire markers. These markers were inserted into the spinous processes of the lumbar vertebrae and into the posterior superior spines of the ilium. The change in angles was photographed and measured. Most of the braces had to be modified to some degree to allow postural measurement by the wire method.

Results indicate the type of brace used is a significant factor in lumbosacral spinal posture. The authors hypothesized that small changes at the lumbosacral level would have a large affect on the trunk and upper extremities of the body. Another large effect was the participants flexion/extension ability. One participant had poor ability to bend forward and subsequently exhibited much more flexion in his lower lumbar region of the spine than one who had greater flexion abilities. It is not clear from this study how a lifting task involving an external load would affect observed results.

2.5.2. Million, Nilsen, Jayson, and Baker (1981)

Spinal motion was measured and a back pain questionnaire was administered to help describe the role of lumbar support corsets in relief of back pain. Nineteen patients male and female patients who had suffered from chronic back pain for at least 6 months and who had not responded satisfactorily to any form of treatment participated in the study. Each patient was randomly assigned a lumbar corset with or without a spinal insert. There was not an unsupported condition. It was hypothesized that the relief of symptoms of back pain may be related to the spinal support restricting movements and making the subject sit and stand in a "better" posture, or alternatively increasing IAP. The subjective questionnaire was directed at establishing the severity of symptoms and how those symptoms interfered with normal activities. Objective measurements of posture, through the use of a tape measure and a goniometer, included straight leg extension and lumbar movement (Table 1). Assessments were made at the beginning of the study and after 4 and 8 weeks of corset use.

Significant improvement based upon subjective responses was observed over the testing period for those who had the support as compared to those who had no support (Figure 4). Objective measurements (Figure 5) indicated there were no significant differences between the 2 groups over the 10 measurements for the duration of the study. Million et al. (1981) suggested the principal method of relief of symptoms was the restriction of spinal motion that the lumbar support imposed.

Table 1. Objective assessments (Million et al., 1981).

1. Straight leg raising - right (degrees)
2. Straight leg raising - left (degrees)
3. Lumbar extension (degrees)
4. Lumbar extension (cm)
5. Lumbar lateral flexion - right (degrees)
6. Lumbar lateral flexion - right (cm)
7. Lumbar lateral flexion - left (degrees)
8. Lumbar lateral flexion - left (cm)
9. Lumbar flexion (degrees)
10. Lumbar flexion (cm)

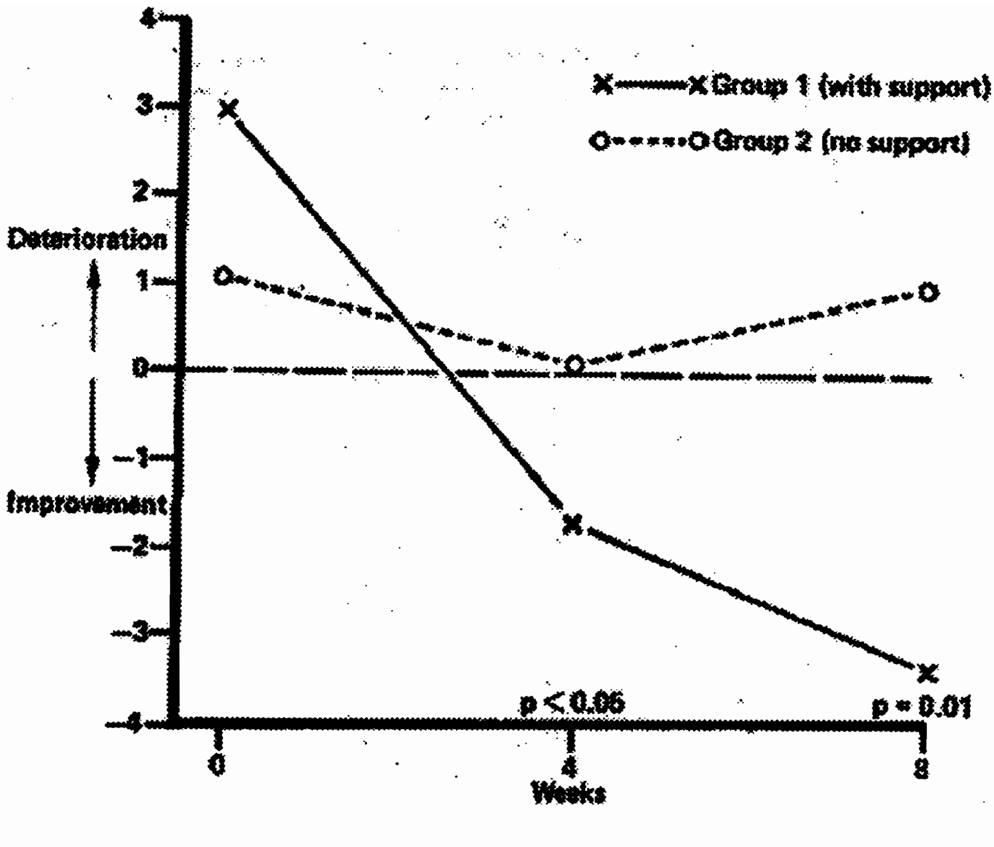

Figure 4. Alteration in subjective index. The improvement in those with the support is significantly greater than in those without the support at 4 and 8 weeks. Taken from Million et al. (1981).

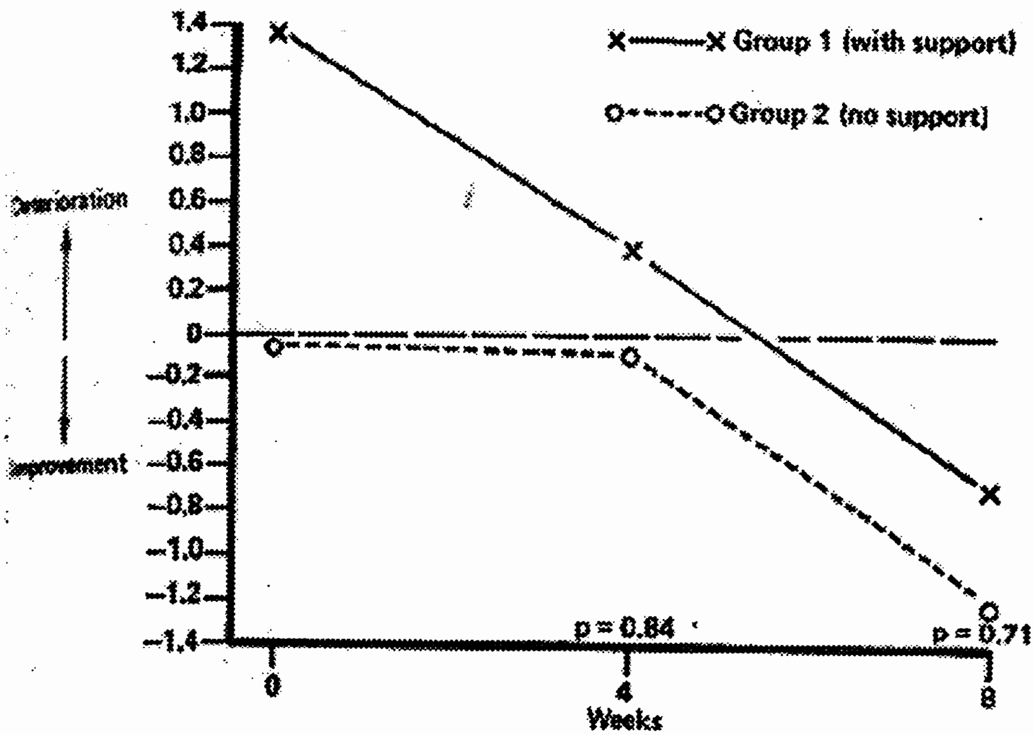

Figure 5. Alterations in objective index. The improvements in the objective index did not differ between the two groups. Taken from Million et al. (1981).

2. 5.3. Grew and Deane (1982)

The physical effects of lumbar spinal supports with regard to their effect upon spinal movements, skin temperature, and IAP were of interest in this study. Eight patients who suffered from chronic low-back pain and 10 healthy volunteers participated in the experiment. Their symptoms were in remission and they had been wearing lumbar spinal supports regularly for at least three months. Participants were required to perform a sequence of spinal movements. Each sequence was performed six times, once without a support and then repeated five times while wearing five different abdominal support braces (Table 2). Spinal movements of the lumbar region were recorded using a vector stereograph. To isolate lumbar movements, a constraint frame was constructed to eliminate pelvis movements. All movements (Table 3) were measured to the limit of subjective comfort.

Results of this investigation concluded spinal supports influence movements significantly as well as skin temperature and IAP. Grew and Deane (1982) hypothesize the presence of the support reduces the need for activity of the muscles of the abdominal wall. Reduced muscle activity was assumed due to the increase in IAP.

2.5.4. Fidler and Plasmans (1983)

The objective of this investigation was to assess and compare the effects of the lumbosacral corset, the Raney flexion jacket, the Baycast jacket, and the Baycast jacket with inclusion of the left thigh (Baycast Spica) on the segmental sagittal mobility of the lumbosacral spine. Five males participated in the experiment. Lateral view radiographs of the lumbosacral spine were made

Table 2. Tested supports (Grew and Deane, 1982).

Symbol	Description
NS	No support.
SE	Semi-elasticated, narrow corset. Padded lumbar insert semi-conforming to lordosis. Rigid anterior section.
NF	Narrow fabric corset with some posterior strengthening.
LF	Long fabric corset extending from pelvis to thorax. Some steel posterior strengthening and some padding.
RB	Leather covered steel brace. Pelvic and thoracic hoops linked by longitudinal members. Anterior abdominal pad.
PJ	Polythene jacket.

Table 3. Spinal movement instructions (Grew and Deane, 1982).

Terminology	Instruction
Neutral	Stand upright comfortably
Flexion	Flex fully forward
Circumduction	Move to your left rotating forward-left, left, backward-left, back backward-right, right, forward-right, forward.
Extension	Lean back
L. lateral bend	Lean left
R. lateral bend	Lean right
Circumduction	Flex forward and then move to your right rotating forward-right, right, etc. to forward position

during maximal flexion and extension. Similar to the procedure for roentgenograms used by Norton and Brown, (1957), lines were drawn on the radiographs along the end-plates of each vertebra. The angles between adjacent vertebrae allowed measurement of spinal angular motion.

Results indicated that almost all of the supports significantly reduced the segmental angular movements of the lumbosacral spine compared to the non-supported control condition (Figure 6). Considerable variation was observed due to the experimental design; each volunteer acted as his own control.

2.5.5. McCoy (1986)

McCoy (1986) investigated the pyschophysical, physical, and psychological effects of the use of two different commercially available support belts on the user. The experiment consisted of a psychophysical lifting task followed by a subjective survey. A tote box containing steel shot was lifted at a rate of three lifts per minute for 45 minutes from the floor to the metacarpal III height (NASA, 1978). The box was automatically lowered to the floor after each lift by a specially constructed device. Abdominal pressure changes were estimated by recording the external pressure between the support belt and the abdomen with a Texas Instruments Pressure Evaluator. McCoy (1986) established the criteria that a significant change in perceived weight from the control situation indicated a significant effect due to abdominal support belt use. Twelve males participated in the experiment.

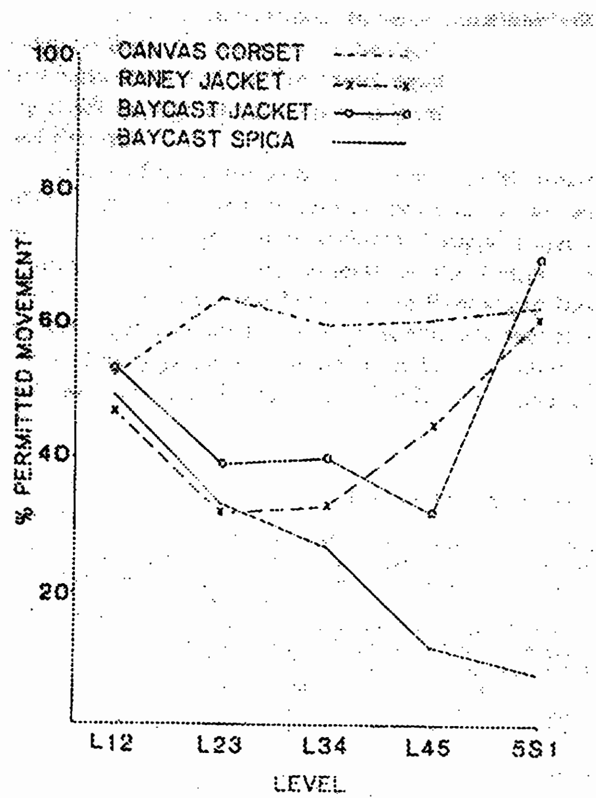

Figure 6. Percent of movement permitted at each level by each type of support. Taken from Fidler and Plasmans (1983).

McCoy (1986) found that both support devices increased the perceived maximum acceptable weight of lift significantly over the control weight. With regards to external pressure measurements, no significant differences were found between either support device. In general survey results indicated that one-third of the participants preferred to work with no belt. Reasons generally centered on comfort issues rather than performance of the supports. Unlike the studies performed by Wilson (1989) and Horsford (1989), the subjective survey revealed that participants found the belts not restrictive with regards to range-of-motion. What separated this study from those performed by Wilson (1989) and Horsford (1989) was the psychophysical lifting task investigated 50% and 75% maximum voluntary lift capability at one lift per minute for ten minutes versus 100% maximum voluntary lift capability at three lifts per minute for forty-five minutes.

2.5.6. Lantz and Schultz (1986)

The effects of wearing commonly prescribed low-back braces and corsets on restriction of body motions were investigated. Three different supports were examined: a lumbosacral corset (LSC), a chairback brace (CBC), and a molded plastic thoracolumbosacral orthosis (TLSO). Five males participated in the investigation. Balsa wood rods were fastened over the abdomen and on the back at the level of the L3 vertebra. In addition, rods that extended laterally were fastened to the shoulders and a marker was placed over the xiphoid process. Movement in three planes was recorded by cameras placed above, to the left, and to the front of the participant each quantifying twisting, flexion/extension, and lateral bending motions, respectively. Participants performed 35 tasks in which they were asked to move maximally. Movements included flexion, extension, right and left twisting, and right and left lateral bending. Combinations of the above movements were also performed. Percent restriction was then calculated by:

$$\text{percent restriction} = \frac{(\text{no-orthosis value}) - (\text{orthosis value})}{(\text{no-orthosis value})} \times 100 \qquad (2.1),$$

The calculations were averaged across participants.

Results indicate that all three supports significantly restricted some of the body motions examined; none permitted any notable mean increases over the control condition. It was found for the various forms of maximal motion, the TLSO was most effective in restricting body motion and the LSC usually the least effective (Table 4). From these results, Lantz and Schultz (1986) concluded that performance of daily activities is likely affected due to support usage. How much performance is affected is difficult to determine due to the fact the most people will not perform daily activities, such as lifting or reaching, at their maximal range limit of motion in a majority of cases.

2.5.7. Amendola (1988)

Amendola (1988) wanted to assess the utility of the use of support belts for manual lifting. As in the previous study by McCoy (1986), the same two commercially available abdominal support devices (Air Belt and CompVest) were tested in addition to a third belt which was a combination of the characteristics of the two belts tested. Twelve males participated in this experiment. The participants were required to lift a tote box filled with steel

Table 4. Percent mean restriction (Lantz and Schultz, 1986).

Motion	LSC	CBB	TLSO
Standing			
twisting §	17	36	46
lateral bending §	22	37	42
flexion	15	9	8
extension	9	12 ‡	23
Sitting			
twisting §	-2	16	27
lateral bending §	29	45	48
flexion	1	3	20
extension ¶	28	27	48
Mean over the eight modes			
LSC		15	
CBB		23	
TLSO		32	

§ includes both right and left motions
‡ two large increases in motion skewed the mean value for the group
¶ with back support

shot to the metacarpal III height (NASA, 1978). Two different loads were lifted, light and heavy, as determine by a psychophysical technique. Frequency of lift was varied from one to three lifts per minute. Amendola (1988) collected data by means of both psychophysical information and subjective surveys. Table 5 presents the questions asked in the subjective survey. In addition, posture data taken from video using reflective markers was also recorded. For the biomechanical analysis, the ankle, knee, hip, shoulder, elbow and wrist joint-centers were digitized. A biomechanical model program developed by Chaffin (1969) was used to determine low-back compressive forces at the L5/S1 disc.

In contrast to the findings of McCoy (1986), it was found that all three abdominal support devices did not increase the maximum acceptable weight of lift. Only the frequency of lift was found to be significant. Among the different subjective questions surveyed, participants had reported a significant amount of restricted motion as a result of belt use. Biomechanical analysis indicated no significant differences in compression between the support belts and the control for either load lifted. A significant difference in compression was found between the different loads. Lifting technique was not controlled, therefore it is very difficult to determine if posture was affected across trials. A posture effect does, however, appear to be present according to the subjective surveys. Participants felt their posture was altered as a result of belt use.

Table 5. Participant survey questions (Amendola, 1986).

1. How would you rate the help given by the belt?

1------------2--------------3-------------4-------------5

no help   some   very helpful

2. Did you feel the belt was too hot?

1------------2--------------3-------------4-------------5

no   slightly   very hot

3. Was your blood circulation restricted by the belt?

1------------2--------------3-------------4-------------5

no   mild discomfort   numbness

4. Were adjustments necessary to make the belt comfortable?

1------------2--------------3-------------4-------------5

no adjust.   some adjust.   many adjust.

5. Did the belt restrict your freedom of movement?

1------------2--------------3-------------4-------------5

no   some   very restrictive

6. If your job required materials handling and provided a belt for you, would you wear it?

1------------2--------------3-------------4-------------5

no   possibly   yes

2.5.8. Sherwood (1988)

Identical to the method used by Amendola (1988) on males, Sherwood (1988) evaluated back support devices for females. The same experimental procedures, data collection methods, and metric were adopted as in the Amendola (1988) study with the only difference being in the frequency of lift. Frequency of lift was set at three lifts per minute. Identical results were observed by Sherwood (1988) as found by Amendola (1988) with males. Participants felt range-of-motion was restricted and biomechanical analysis revealed no difference between belt/nonbelt conditions.

2.5.9. Lander, Simonton, and Giacobbe (1989)

The purpose of this study was to examine the effectiveness of weight-belts during the performance of the parallel squat exercise. Trials consisted of three Belt conditions of none, light, and heavy with three load conditions of 70, 80, and 90% 1RM (one repetition maximum). Six skilled males trained in weightlifting techniques served as participants. The light weight belt (used by most recreational weight lifters) consisted of a single layer of leather 7 mm thick and 100 mm wide in the center which tapered to 73 mm at both ends near the buckle. The heavy weight belt (used by most competitive power lifters) consisted of three layers of leather 11 mm thick and 100 mm wide for its entire length. A number of different types of data were collected during the experiment. A high speed camera, placed perpendicular to the sagittal plane of motion, was used to film each trial for posture digitization. A force platform was placed in the center of a lifting platform. A pressure balloon catheter was inserted into each participants' rectum in order to collect IAP readings. In addition, EMG signals from the rectus abdominus, external obliques, and lower erector spinae muscle groups were also collected. A quasi-static model was then implemented to predict compressive, shear, and muscle force values acting at the L5/S1 intervertebral disc.

The significant results were observed when the heaviest weights were lifted and consequently were the only results reported. No significant differences were observed between conditions for any of the absolute and relative joint angles examined. IAP was greatest for the light and heavy Belt conditions and least for the nonbelt condition. From EMG data it was observed that a smaller amount of muscle activity was present when either type of belt was worn. From this investigation it appears that posture was not effected as a result of belt usage. It is not clear how past experience and skill in performing this particular lift affected posture.

2.5.10. Wilson (1989)

Wilson (1989) was interested in assessing the utility of the use of three different external abdominal support belts for female users. The two support devices tested were commercially available abdominal support belts while the third was a commonly used weight lifting training belt. The two commercial belts were manufactured by Air Belt and CompVest, both used in the studies performed by McCoy (1986), Amendola (1988) and Sherwood (1988). Two levels of maximum voluntary lift capability were investigated (50 and 75%). A psychophysical lifting experiment was conducted with the frequency of lift set at one lift per minute. Each trial lasted for ten minutes. Twelve healthy female participants were evaluated by psychophysical, electromyographical

(EMG), biomechanical, subjective survey (rating and ranking), and body part discomfort methods. Table 4 presents the questions asked in the subjective survey. Each of the participants lifted a tote box containing steel shot from the floor to the metacarpal III height (NASA, 1978) while wearing the three different support devices. Erector spinae muscle activity was recorded through EMG measurement. Each participant was video taped with reflective markers attached at different joint centers for later digitization and analysis. Software developed by Congleton, Foster, Fleischer, and Villalobos (1989) titled *Video Ergonomic Evaluator* was used for the biomechanical analysis to calculate forces on the lower back at the L5/S1 disc. The biomechanical model used by this software was developed by Chaffin and Andersen (1984).

In general, the three back support devices did not significantly differ with regards to preference. However participants reported the belts restricted freedom of movement to a small degree. The leather belt was found to be significantly more restrictive than the Air Belt and CompVest devices at the 50% exertion level while no change in level of restrictiveness was observed at the 75% exertion level. Biomechanical results indicated compression at the L5/S1 disc and shear at the L5/S1 were not significantly different for either load percentage across the belt/nonbelt conditions. Erector spinae muscle force was found to be significant across exertion levels. Although these results indicate postural changes did not occur as a result of belt use, it is difficult to support this conclusion outright. Joint center (marker location) differences were not compared across testing conditions.

2.5.11. Horsford (1989)

Virtually identical to the research performed by Wilson (1989) on females, Horsford (1989) investigated the effectiveness of back support belts in males. The same support belts types were tested to determine which device resulted in less EMG-measured muscle activity and subjective discomfort. Twelve male participants took part in this study. In addition, Horsford (1989) was interested in determining the relative biomechanical forces imposed in the L5/S1 region as a result of belt usage. As used by Wilson (1989), the biomechanical analysis was performed using the video digitization technique developed by Congleton, Foster, Fleischer, and Villalobos (1989) and the biomechanical model developed by Chaffin and Andersen (1984).

Results were identical to those obtained by Wilson (1989) for females. Participants reported belt use resulted in a restricted range-of-motion. Compressive and shear forces at the L5/S1 disc were not significant across belt/nonbelt conditions. Predicted erector spinae muscle force was larger for the higher percentage exertions. Horsford (1989) recommended, like Wilson (1989), that future experiments should be expanded to encompass the entire lifting cycle including asymmetric lifts, awkward lifting, pushing, and pulling activities.

2.5.12. Reddell (1992)

This clinical study looked at the efficacy of a commercially available support belt in relation to the reduction of the lumbar injury incident rate and the severity of injuries over an eight month period of time. The participants were 642 baggage handlers working for a major airline company.

Data collected included subjective questionnaire answers (virtually identical to the one used by McCoy (1986), Amendola (1988), Sherwood (1988), Wilson (1989), and Horsford (1989)), total number of lumbar injuries, number of lost work day case lumbar injuries, number of restricted work day case lumbar injuries, lost work days, restricted work days, total worker's compensation cost, and number of hours worked for the duration of the study. The fleet baggage handlers worked in five different general locations for the airline company: inside/outside aircraft, baggage transfer, mail facility, cabin service, and bag room operations. All tasks performed at each of these work areas included manual transfer of baggage, mail, or supplies from one location to another. Participants were randomly assigned to four different treatment groups: participants who received a support belt, participants who received training class on lifting techniques, participants who received both a support belt and training class on lifting techniques, and a control group.

During the first session, anthropometric data was recorded and support belts and lift training class booklets were assigned. At the end of eight months, each participant was interviewed and questionnaires were completed. Results indicated that compliance was a major problem with regard to support belt use. Fifty-eight percent of the participants who were issued a belt discontinued its use prior to the end of the eight month period. Of those who continued use of the belt, 26% indicated that the belt rubbed, pinched, and bruised ribs. In addition, 9% felt the belt was uncomfortable and 8% percent reported the belt was too restrictive of body motion. Although difficult to determine from this research, posture interference due to belt use during daily activities appears to be evident.

2.5.13. Other Recent Studies Involving Postural Aspects and Support Belt Use

McGill, Seguin, and Bennett (1993) tested flexibility and stiffness of the lumbar torsos of 20 male and 15 female participants, while wearing and not wearing a leather abdominal support belt. It was observed that the stiffness of the torso was significantly increased about the lateral bend and axial twist axes, but not when participants were rotated into full flexion. McGill et. al (1993) concluded that abdominal belts assist to restrict the range-of-motion about the lateral bend and axial twist axes but do not have the same effect when the torso is forced in flexion. In essence, a difference in posture is more likely during an asymmetric lift than when a symmetric lift is performed.

According to Adams and Hutton (1988) the compressive strength of the lumbar spine decreases when the end range-of-motion in flexion is approached. These observations would lead one to expect the risk of injury to decreased due to the restriction of not allowing the body to assume a posture near the end range-of-motion during asymmetric lifting situations when a support belt is worn. This aspect, however, has not been objectively tested.

2.6. The Inclusion of IAP Effects with Regard to Low-Back Modelling

A number of different low-back biomechanical models have been developed to investigate low-back issues ranging from the prediction of maximum allowable loads in various postures to the percent activation of certain muscle groups within the trunk area. With regards to IAP, it has been vigorously debated as to whether biomechanical models of the low-back should incorporate IAP as part of their construction. Of more importance to this research is the possibility of an increase in IAP without an increase in trunk muscular activity as a result of abdominal support belt usage. The question of whether this aspect should be included in biomechanical models of the low-back remains.

In a study performed by McGill, Norman, and Sharratt (1990), they investigated the contribution (if any) to IAP generation by the use of abdominal support belts. They specifically investigated whether abdominal support belts reduced trunk muscle activity and/or increased IAP. Six participants lifted loads with and without wearing a competition weightlifter belt. In addition, the effects of breath holding and exhalation were investigated during the experiment. McGill et al. (1990) found that the use of the competition lifting belt slightly increased IAP but no appreciable difference in rectus abdominis or erector spinae EMG activity was observed. They concluded support belts do not appear to contribute to supporting the loaded lumbar spine. However, it was found that holding one's breath causes an increased IAP and tends to reduce erector spinae activity whether a belt is worn or not worn.

As discussed previously, Harman et al. (1989) concluded that a lifting belt augments IAP during lifting exercises. It was also hypothesized that IAP is similarly affected by a belt during other lifts involving back extension against resistance. However, unlike the McGill et al. (1990) study, Harman et al. (1989) did not record EMG activity which might have provided information regarding muscle activity during both belt and nonbelt trials.

Other studies have been performed which raise questions on how IAP relates to EMG muscle activity and posture. Mairiaux and Malchaire (1988) found differences in IAP depending on whether the trunk was flexed or extended. They also found that no significant relation between IAP variations and lumbar moments existed when lifting was carried out from a flexed trunk position. Chaffin and Andersson (1991) noted that asymmetrical trunk loading creates large trunk stresses without the corresponding IAP response that occurs in symmetrical loading. Hemborg and Moritz (1985) reported approximately the same IAP values during lifting in low-back patients and in healthy controls despite a significant difference in abdominal muscle strength between the two groups.

What becomes very evident is that the role of IAP in biomechanical low-back modeling is not yet completely understood. An increase in IAP (if any) may not be a result of belt usage but possibly due to an increase in the activity of the muscle pairs surrounding the trunk or simply the result of breath holding as demonstrated by McGill et al. (1990).

2.7. Optimization-Based Biomechanical Modeling

Chaffin and Andersson (1991) describe biomechanical models as mechanical representations of the operation of the musculoskeletal system of the body. These representations involve the application of engineering concepts and the laws of physics to describe forces and motions acting on various body parts. As discussed by Johnson (1992), the need for low-back biomechanical models is motivated by three concerns. First, biomechanical models allow interpretation of the large amounts of data available from advanced bioinstrumentation. Second, biomechanical models make it possible to assess the possible risks of some tasks without purposely inducing injury to humans and/or human tissue. Third, biomechanical models can be used to assess certain types of tasks that could not otherwise be practically measured.

Static biomechanical modeling, as used in this research, require that all forces and moments along each axis must sum to zero

$$\begin{aligned}\sum \mathbf{F}_{\text{external}} + \sum \mathbf{F}_{\text{internal}} &= 0 \\ \sum \mathbf{M}_{\text{external}} + \sum \mathbf{M}_{\text{internal}} &= 0\end{aligned} \qquad (2.2).$$

If these above equations are not satisfied, the system is not static. If the external reaction is composed of more unknown muscle forces than the number of equilibrium equations, the solution is indeterminate. This is the case for modeling of the trunk and torso region of the body. As discussed in section 2.2., the trunk and torso is made up of five main muscle pairs. One method that can be used to solve this indeterminate system is through the use of optimization techniques.

The optimization technique used in the biomechanical model for this research involves the use of a double linear programming method. In an algorithm proposed by Bean, Chaffin, and Shultz (1988), the technique solves a two-objective problem with two successive linear programs. The first objective is to minimize maximum muscle contraction intensity. The second objective is to minimize the sum of the muscle forces using the intensities found from the first objective:

First Objective - Linear Program

$$\text{minimize } I \qquad (2.3.),$$

$$\text{subject to:}\quad \sum_{i=1}^{n} \|\mathbf{f}_i\| (\mathbf{r}_i \times \mathbf{t}_i) = -\mathbf{M}_{ext}$$

$$\frac{\|\mathbf{f}_i\|}{A_i} \leq I,$$

$$\|\mathbf{f}_i\| \geq 0,$$

Second Objective - Linear Program

$$\text{minimize} \sum_{i=1}^{n} \|\mathbf{f}_i\| t^i_z \qquad (2.4.),$$

$$\text{subject to:}\quad \sum_{i=1}^{n} \|\mathbf{f}_i\| (\mathbf{r}_i \times \mathbf{t}_i) = -\mathbf{M}_{ext}$$

$$\frac{\|\mathbf{f}_i\|}{A_i} \leq I^*,$$

$$\|\mathbf{f}_i\| \geq 0,$$

Optimization-based biomechanical models are usually validated through electromyographical studies. Hughes (1991) examined the Bean et al. (1988) model (Minimum Intensity Compression or MIC) along with another model through electromyographical techniques. It was found that the MIC failed to predict differential levels of erector spinae force during lateral bending motion. This model also does not take into account antagonistic muscle activity or IAP effects.

Although optimization models, such as the MIC, have their shortcomings, these models provide far better results than EMG-based low-back models. EMG models require substantial instrumentation and are quite intrusive, limiting the extent to which they can be applied in a practical setting. With regard to the MIC model, Bean et al. (1988) point out that this model has the advantages of low cost, stable solutions, provides insight into solution sensitivity, and efficient to implement on personal microcomputers. It is for this reason the MIC model was used to predict compression in this research. The MIC model is presented in Appendix C.

2.8. Commercially Available Support Belts

A variety of abdominal support belts are available on the commercial market. A vast majority of these belts have a double-layered construction. The first layer fits around the waist and is held into place by velcro. This layer usually has either stiff plastic, rubber, or steel strips sewn into the back of the belt which provide support to the lumbar region. The second layer is usually made of an elastic material which can be stretched around the waist and secured on the front again by velcro. The stretching of the second layer into place secures the belt tightly around the waist and lumbar area. The width of this type of belt at the lumbar region, depending on the type of model and manufacturer, usually varies between 8 and 10 inches. Differences across models and manufacturers include the inclusion of shoulder straps, apron and utility tool storage attachments, color, and slight differences in belt cut. This design is incorporated into the majority of belts produced by support belt manufacturers. It is for this reason a belt of this design was used for this investigation.

Other support belts less used in industry include an orthodox leather weightlifting belt. This type of belt is adjusted by a buckle in the front. The width of belt in the lumbar area usually ranges between 4 and 6 inches. A third type of belt mimics the leather weightlifting belt except it is usually made of a foam fabric. It is secured in the front by means of velcro and its width around the lumbar region ranges between 4 to 6 inches. Variations include those which have rigid inserts and air bladders which can be inflated which provide additional support. These belts have been reported to be warmer when worn as compared to the other belt types.

2.9. Literature Summary

As evident from the literature review, it is highly contested whether or not increased IAP exists as a result of abdominal support belt use. What is evident is the influence, whether it be physical or mental, of the use of support belts on lifting posture. According to previous research, lifting posture appears to be significantly affected by the wearing of a support belt, the characteristics of the external load involved, and the nature of the lifting task motion (symmetric or asymmetric).

It was the intent of this thesis to investigate the effects of the use of a commercially available abdominal support belt on torso posture, static lift strength generation, and predicted compression at the L3/L4 intervertebral disc. It was hypothesized that the use of an abdominal support belt will have a significant influence on torso posture especially for those lifts that involve heavy weights and are initiated at the extreme ranges of motion (i.e. asymmetric low lifts). The factors of interest are defined in the Method section. Table 6 provides a summary of the literature regarding abdominal support devices as it applies to this investigation.

Table 6. Literature summary.

Author	Results
Waters & Morris (1970)	• measured effect on EMG values believed due to phys. const. of belt • lower extremity function not affected by belt use
Kumar & Godfrey (1986)	• no differences in IAP observed across different belt types
Harmon et al. (1989)	• IAP rose earlier during lifting when a belt was worn • average IAP rise slower when a belt was worn
Hunter et al. (1989)	• measured blood pressure and heart rate significantly higher
Lander et al. (1990)	• belt repetitions performed faster than nonbelt repetitions • IAP values were 25-40% higher when a belt was worn
Walsh & Schwartz (1990)	• combination of "back school" and support belt usage resulted in a significantly less amount of work days missed • no significant change in abdominal strength after prolonged use of support belts between groups after duration of 6 months
McGill at al. (1990)	• breath holding appears to unload spine, not affected by belt use
Norton & Brown (1957)	• differences observed at adjacent lumbar interspaces for each brace
Million et al. (1981)	• no change in range-of-motion and subjective improvement observed
Grew & Dean (1982)	• range-of-motion influenced significantly due to belt use
Fidler & Plasmans (1983)	• significant reduction of segmental vertebral angles, each tested support
McCoy (1986)	• no subjective change in restrictiveness of belts • significant change in weight lifted as a result of belt use
Lantz & Schultz (1986)	• significant reduction in body motions observed for all supports
Amendola (1988)	• frequency of lift and subjective reporting of restrictiveness signif.
Sherwood (1988)	• frequency of lift and subjective reporting of restrictiveness signif.
Lander et al. (1988)	• no change in joint angles observed as a result of belt use
Wilson (1989)	• subjective reporting of restrictive movement, biomech. inconclusive
Horsford (1989)	• subjective reporting of restrictive movement, biomech. inconclusive
Reddell (1992)	• compliance problem, discomfort and restricted motion reported

3. EXPERIMENTAL METHOD

This chapter describes the method used to determine if a change in torso posture occurred as a result of support belt use across a series of static lift trials. Participants were asked to perform a number of static maximal lifts, both initiated from symmetric and asymmetric positions differing in height, while using and not using an abdominal support belt. Their posture and reactive forces at the hands were recorded. The posture data of the nonbelt condition was then compared with the observed postures for the Belt conditions. In addition, strength exertion values and spinal compression values computed at the L3/L4 intervertebral disc for the different conditions were also compared.

3.1. Participants

Sixteen college students (8 female and 8 male) served as participants in this research. They were recruited from the student body at Virginia Tech by means of advertisement. Only those subjects who were in good health and not had a history of any low-back pain or musculoskeletal disorders were allowed to participate. Each participant was reimbursed $5 per hour for their time. Table 7 shows the anthropometric statistics collected from the participant group.

3.2. Apparatus

The equipment used for this research included a posture recording system, a force platform, a lift rig apparatus, and a commercially available abdominal support belt (in various sizes).

Table 7. Collected anthropometric statistics.

Measure		Males	Females
Age (years)			
	Mean (Std.)	23.63 (2.56)	22.38 (4.27)
	Minimum	21	18
	Maximum	28	32
Stature (cm)			
	Mean (Std.)	181.48 (4.67)	164.58 (7.88)
	Minimum	174.30	157.90
	Maximum	188.30	180.50
Weight (kg)			
	Mean (Std.)	81.76 (17.65)	59.61 (10.34)
	Minimum	54.09	49.40
	Maximum	109.09	83.18
Bicrestal Breadth (cm)			
	Mean (Std.)	29.35 (3.27)	28.00 (3.11)
	Minimum	24.40	24.10
	Maximum	34.20	34.00
Trochanteric Height (cm)			
	Mean (Std.)	94.20 (4.64)	84.36 (4.82)
	Minimum	87.10	78.20
	Maximum	99.90	94.10
Vertebra Prominens Depth (cm)			
	Mean (Std.)	25.28 (1.70)	24.38 (1.64)
	Minimum	22.70	21.80
	Maximum	27.00	26.10
Standing Elbow Height (cm)			
	Mean (Std.)	116.65 (3.55)	105.45 (6.24)
	Minimum	109.20	99.60
	Maximum	119.80	117.20
Calf Height (cm)			
	Mean (Std.)	35.51 (1.74)	31.35 (1.93)
	Minimum	32.80	27.90
	Maximum	37.50	33.90

3.2.1. Posture Measurement System

A Northern Digital WATSMART three-dimensional motion analysis system was used to record each posture during the experimental trials. Components of the system include infrared emitting diodes (IREDs), three infrared cameras, a calibration frame, and the system unit. The system unit interfaced to a GRID 386 personal computer. The WATSMART system tracked infrared diodes which were attached to different body targets. Three different cameras were used to cover the experimental area. Each infrared diode must have been seen by at least two cameras to properly measure its location relative to the calibrated space. The WATSMART software reconstructed three-dimensional coordinates from the collected two-dimensional readings. Collection rate was set at 20 Hz. The cameras were mounted to steel poles that are anchored into circular concrete bases. The WATSMART cameras are extremely sensitive to reflections and therefore, the experimental area was bordered by black non-reflecting foam material hung as drapes on three sides. The average RMS error over the posture data collection phase was 4.88 mm with a standard deviation of 0.295 mm. Minimum recorded RMS error was 4.41 mm while the maximum recorded RMS error was 5.56 mm. Figure 7 depicts the overall layout of the experimental apparatus and the locations of the three infrared cameras.

3.2.2. Force Platform

An AMTI Model OR6-5-1 Biomechanics Force Platform was used to simultaneously measure three force components along the X, Y, and Z-axes and

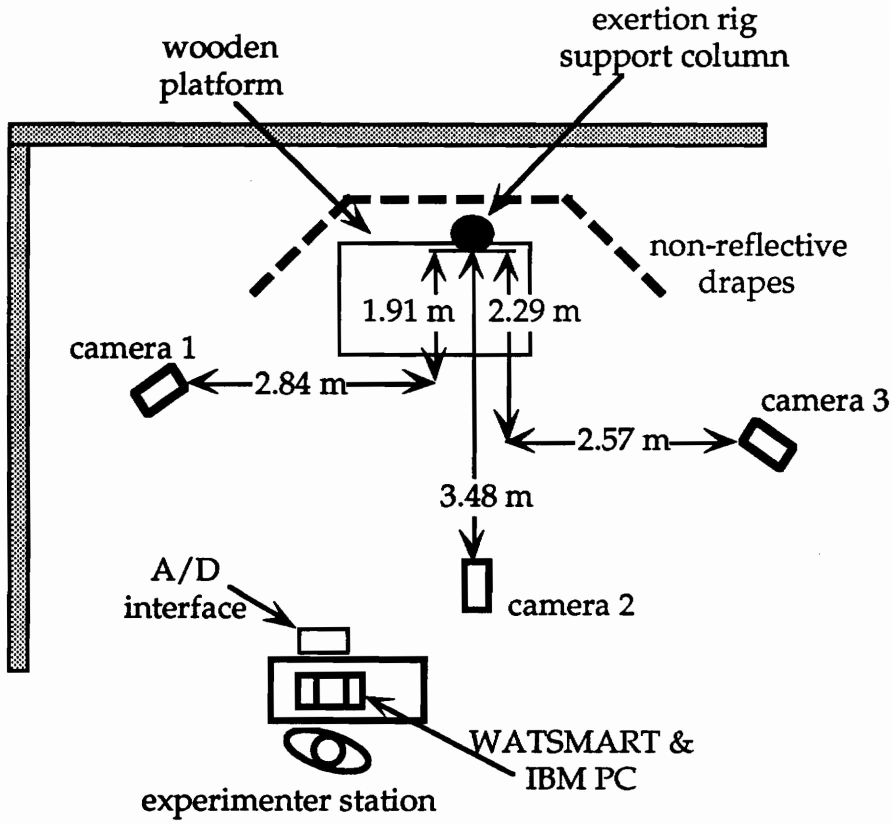

Figure 7. Layout of the experimental apparatus.

three moments about each axis. The platform was connected to an AMTI Model SGA6-4 Strain Gage Amplifier system which in turn was connected to an IBM PS/2 model 50 personal computer. The force platform was used to gather static two-handed force data.

3.2.3. Static Lift Apparatus

A Baltimore Therapeutic Equipment (BTE) column was modified to allow simulation of various static lift tasks. The force platform was attached to the column by a constructed steel frame. A schematic of the rig is shown in Figure 8. An adapter to the force platform simulated the lifting of a box with handles on each side. The handles were oriented parallel to the plane in the floor. The distance between the handles was set at 48 cm. This handle distance is the 50% forearm-forearm breadth as (1965 USAF Survey, NASA 1978). The forearm-forearm breadth is defined as the distance across the tissue mass of the forearms measured with the elbows flexed and resting lightly against the body. Hand forces were measured through the force platform bolted to the steel rig. The BTE column allowed adjustment to the height of the platform in relation to the floor platform. A computer program, written in Pascal, was used to collect force data from the force platform.

3.2.4. Commercially Available Abdominal Support Belt

The abdominal support belt used in this study was the Decade Back Support made by Chase Ergonomics Incorporated (Figures 9 and 10). The belt came in five different sizes depending upon the measured circumference

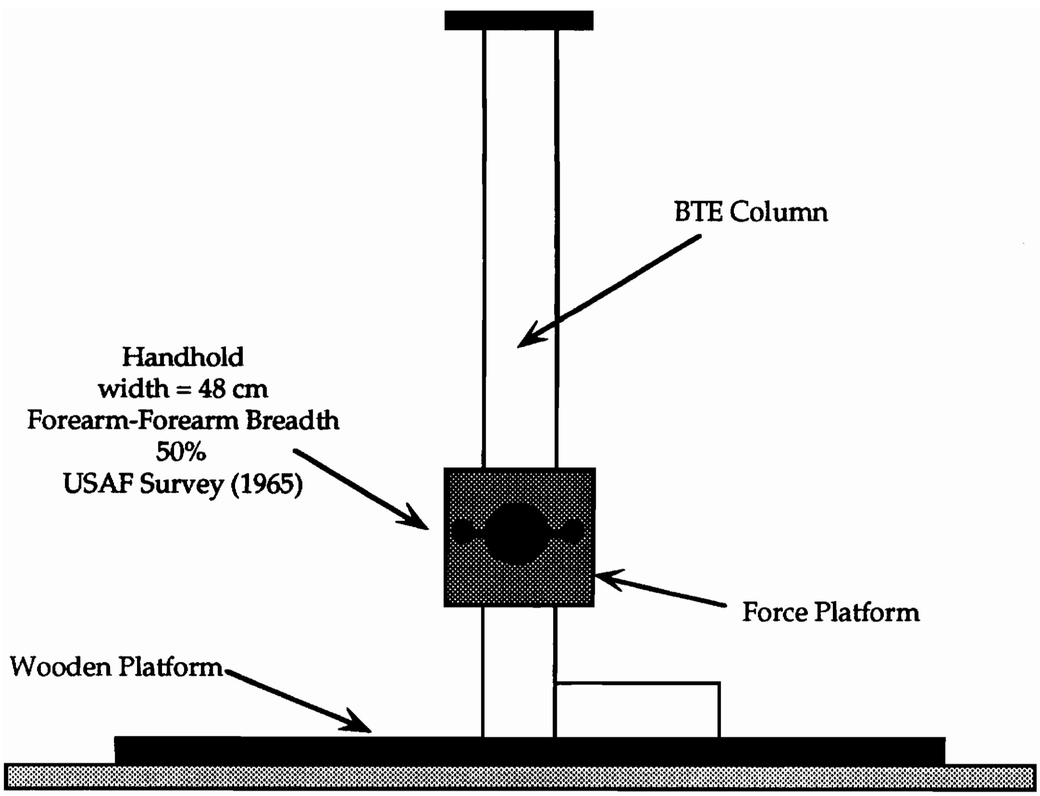

Figure 8. Schematic of the BTE rig.

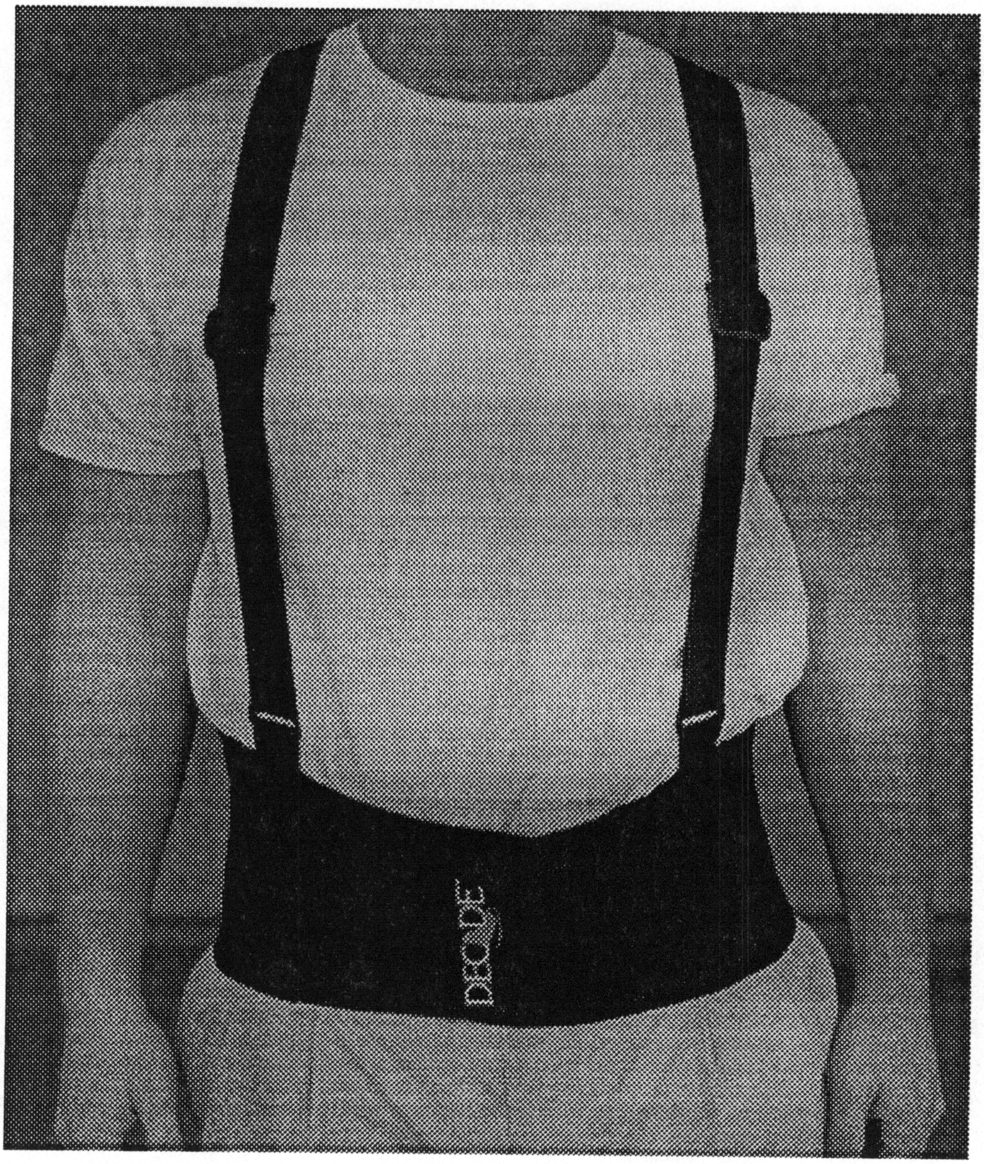

Figure 9. Anterior view - Decade back support.

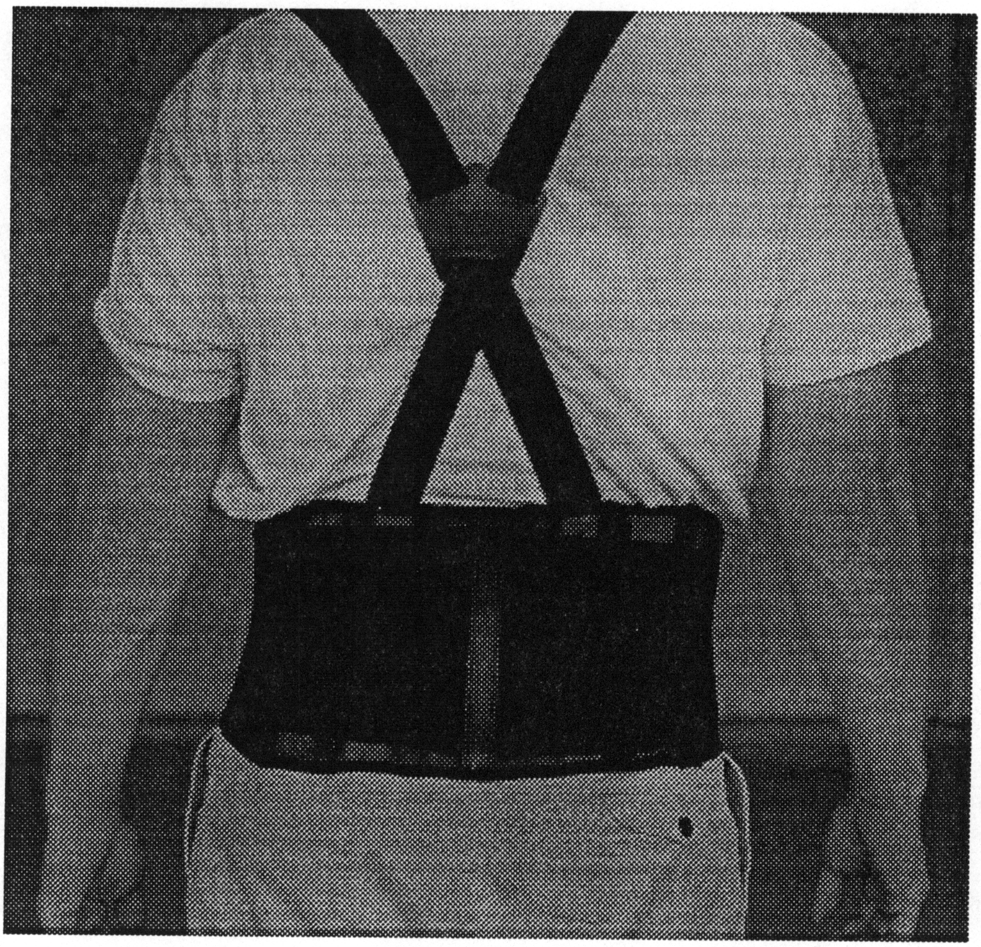

Figure 10. Posterior view - Decade back support.

around the waist of the wearer. This support featured a "double-layered" construction (as described in Section 2.8.) with adjustable shoulder straps. Width of the belt at the back, front, and sides is 8, 6, and 5 inches, respectively. Instructions for the belt's proper use are located in Table 8.

3.3. Marker Placement

Figure 11 describes where the 14 WATSMART markers were placed on the trunk, lower, and upper extremities of the body. IRED markers were placed at the calves of both legs. One marker was placed at the approximate level of the L3/L4 joint-center and another 11 centimeters to the right of the L3/L4 joint-center marker. The remaining markers were used to collect the positions of the left elbow, right elbow, left shoulder, right shoulder, and the C7/T1 (vertebra prominens). These markers were each secured to a stick (of known length) away from each of the above locations. This approach was taken because some of the postures assumed by the participants blocked the view of those particular surface landmarks in at least two cameras. The position of the hands was collected before the trials, but after the experimental space was calibrated. Tape on the lift handles marked where each participant was instructed to grasp the handles (placing the middle finger over the tape) thus controlling hand position.

3.4. Experimental Task

The experimental task consisted of maximal two-handed isometric exertions on the lift handles. All trials were 6-second static exertions where the participant ramped up to their maximum voluntary force in 2 seconds and

Table 8. Decade back support instructions for use.

1) Separate elastic cinch straps (wrap-around straps) from Velcro closure.
2) Lay out so that rubber stiching on vertical stays will be on the inside.
3) Slip on the brace like a vest and pull down, adjusting shoulder straps so that belt will rest low on hips.
4) For proper placement, the bottom of the support should be at the tailbone.
5) Grasp each side of belt and pull forward. Bring the right side past midpoint of abdomen. Bring lift side over right, securing Velcro closure.
6) Grasp free elastic cinch straps and pull both forward at the same time for equal pull on each strap.
7) Press cinch straps into a comfortable position, securing Velcro closure.
8) For the "triple lock option," pull left strap past midpoint. Place right strap over left strap and press to secure Velcro closure.

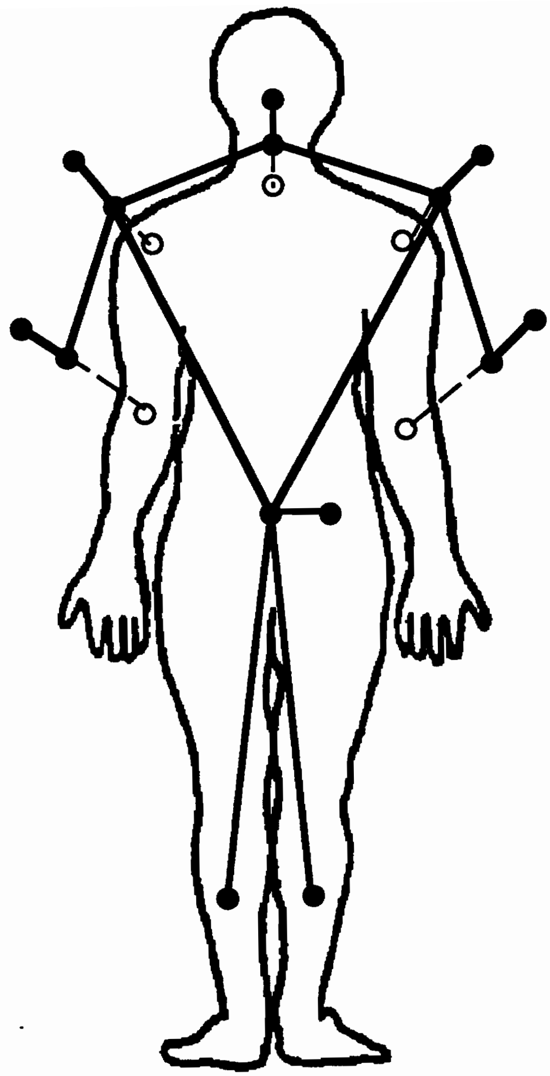

Figure 11. Posterior view of WATSMART IRED diode placement.

then held that force steady for the remaining 4 seconds (Chaffin, 1975). Participants were asked to exhale continuously throughout the exertion so that breath-holding effects were not confounded during the experiment. Hand position was controlled through tape placed on the lift handles. Each participant was instructed to grasp the handles with their middle finger from each hand covering the tape marker during each trial.

3.4.1. Asymmetric Static Lift Conditions

Very similar to the static strength testing performed by Garg and Badger (1986), each asymmetric static lift was set 60° off to the right of the sagittal plane. Each participant was asked to practice each nonbelt asymmetric lift condition before the data collection began. During the practice conditions, foot position measurements were recorded. These positions were then followed during each of the nonbelt and corresponding belt trial conditions. Although participants were required to keep their feet parallel to the sagittal plane, flat on the floor, and in the proper orientation, they were allowed to perform the various lifts using whatever body posture they found to be the most comfortable.

3.4.2. Symmetric Static Lift Conditions

Each symmetric static lift was performed in the sagittal plane. Each participant was asked to practice each symmetric lift condition before the trials began. During the practice conditions, foot position measurements were recorded. These set conditions were then followed during each of the belt and corresponding nonbelt static lift conditions (as described above for the asymmetric lift conditions).

3.5. Experimental Design

The experimental design is a mixed factors design with one between-subjects factor and three within-subjects factors. There were 16 trials, all presented in completely balanced order (Tables 9 through 12) to each of the 16 participants. Each participant performed each static lift condition twice for each Belt condition, Lift type, and handle height.

3.5.1. Independent Variables

Gender: Between subjects, male and female.

Lift type: Within-subjects, two levels: symmetric lift in the sagittal plane and an asymmetric lift 60° to the right from the sagittal plane. It was assumed for this study that there were no differences between asymmetric lifts across the right and left sides of the body. All lifts were two-handed exertions.

Handle height: Within-subjects, two levels: calf height and standing elbow height. Calf height is defined as the vertical distance from the standing surface to the maximum posterior protrusion of the gastrocnemius. Elbow height is defined as the vertical distance from the standing surface to the depression at the elbow between the humerus and the radius. For all measurements, the participant stood erect, arms hanging naturally at his/her sides, heels together, and weight distributed equally on both feet. These measures are described in NASA (1978).

Table 9. Independent variable levels.

Factors	Levels
Lift type	1 - symmetric 2 - asymmetric
Handle height	1 - calf height 2 - standing elbow height
Belt condition	1 - belt 2 - nonbelt

Table 10. Treatment conditions.

Treatment	Lift type	Handle height	Belt condition
1	symmetric	calf	belt
2	asymmetric	calf	belt
3	symmetric	standing elbow	belt
4	asymmetric	standing elbow	belt
5	symmetric	calf	nonbelt
6	asymmetric	calf	nonbelt
7	symmetric	standing elbow	nonbelt
8	asymmetric	standing elbow	nonbelt

Table 11. Treatment presentation order - females.

Participant #1	2	3	4	5	6	7	8	
Treatment	1	2	3	4	5	6	7	8
presentation2	2	3	4	5	6	7	8	1
order	8	1	2	3	4	5	6	7
	3	4	5	6	7	8	1	2
	7	8	1	2	3	4	5	6
	4	5	6	7	8	1	2	3
	6	7	8	1	2	3	4	5
	5	6	7	8	1	2	3	4
	1	2	3	4	5	6	7	8
	2	3	4	5	6	7	8	1
	8	1	2	3	4	5	6	7
	3	4	5	6	7	8	1	2
	7	8	1	2	3	4	5	6
	4	5	6	7	8	1	2	3
	6	7	8	1	2	3	4	5
	5	6	7	8	1	2	3	4

Table 12. Treatment presentation order - males.

Participant #9	10	11	12	13	14	15	16	
Treatment	1	2	3	4	5	6	7	8
presentation2	2	3	4	5	6	7	8	1
order	8	1	2	3	4	5	6	7
	3	4	5	6	7	8	1	2
	7	8	1	2	3	4	5	6
	4	5	6	7	8	1	2	3
	6	7	8	1	2	3	4	5
	5	6	7	8	1	2	3	4
	1	2	3	4	5	6	7	8
	2	3	4	5	6	7	8	1
	8	1	2	3	4	5	6	7
	3	4	5	6	7	8	1	2
	7	8	1	2	3	4	5	6
	4	5	6	7	8	1	2	3
	6	7	8	1	2	3	4	5
	5	6	7	8	1	2	3	4

Belt condition: Within-subjects, two levels: a condition wearing the support belt and a nonbelt condition.

Figures 12 through 19 illustrate the eight different combinations of the independent experimental variables of *Lift type, Handle height,* and *Belt condition.*

3.5.2. Dependent Variables

Three-dimensional torso posture: Upper extremity body posture was recorded with the WATSMART system described in Section 3.2.1 for each trial. Torso posture was expressed with three dependent measures: (1) angle in degrees of extension/flexion, (2) angle in degrees of lateral bend, and (3) angle of axial twist of the shoulders in relation to the hips. All angles were computed with regards to the body reference frame from three-dimensional coordinates (meters) collected during the experimental trials. The vector connecting the L3/L4 joint-center with the vertebra prominens landmark approximated torso aliment. Axial twist angle was defined as the angle between the vector connecting the each shoulder landmark and the vector connecting the L3/L4 joint-center with the lateral L3/L4 marker (which approximated the alignment of the hips).

Two-handed lift force: The three-dimensional force (in N) that participants exerted with both hands on the lift apparatus.

3.6. Experimental Protocol

The experiment took approximately 2.5 hours for each participant. It was conducted in two sessions.

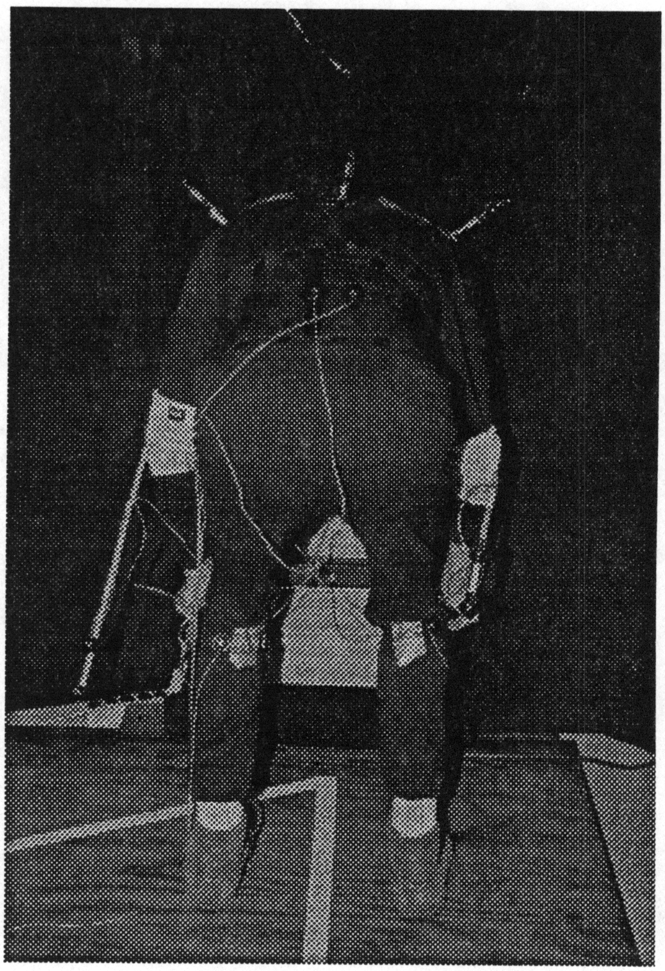

Figure 12. Treatment #1 - symmetric lift, calf height, Belt condition.

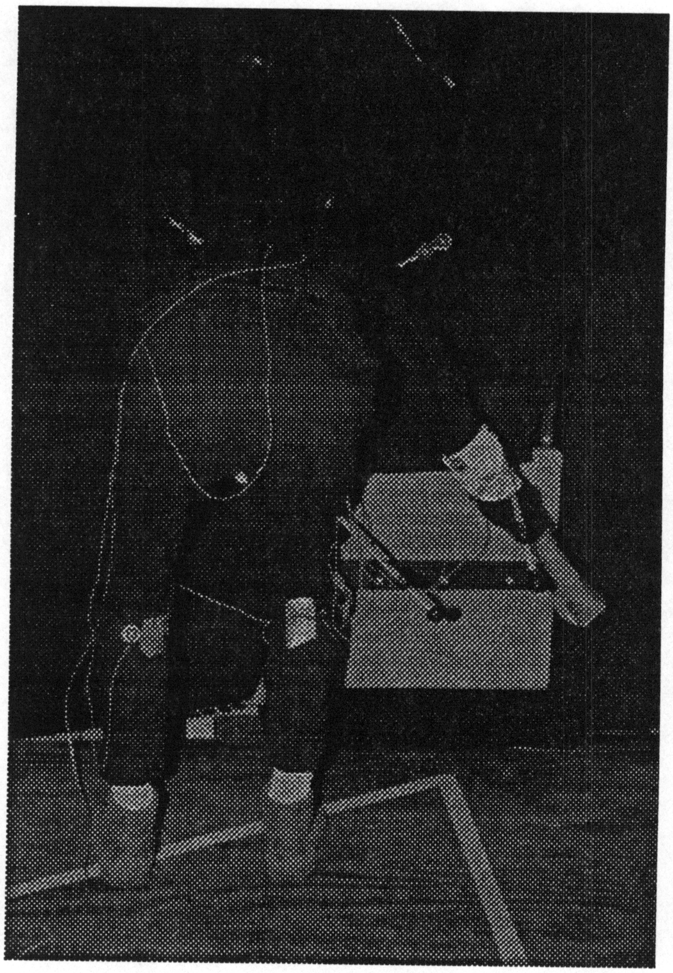

Figure 13. Treatment #2 - asymmetric lift, calf height, Belt condition.

Figure 14. Treatment #3 - symmetric lift, standing elbow height, Belt condition.

Figure 15. Treatment #4 - asymmetric lift, standing elbow height, Belt condition.

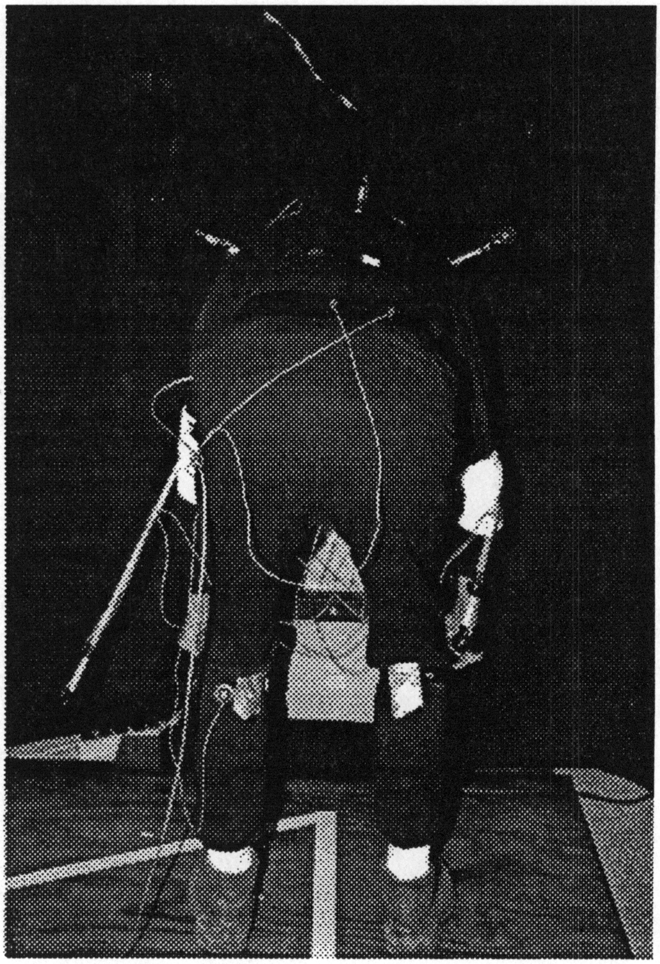

Figure 16. Treatment #5 - symmetric lift, calf height, nonbelt condition.

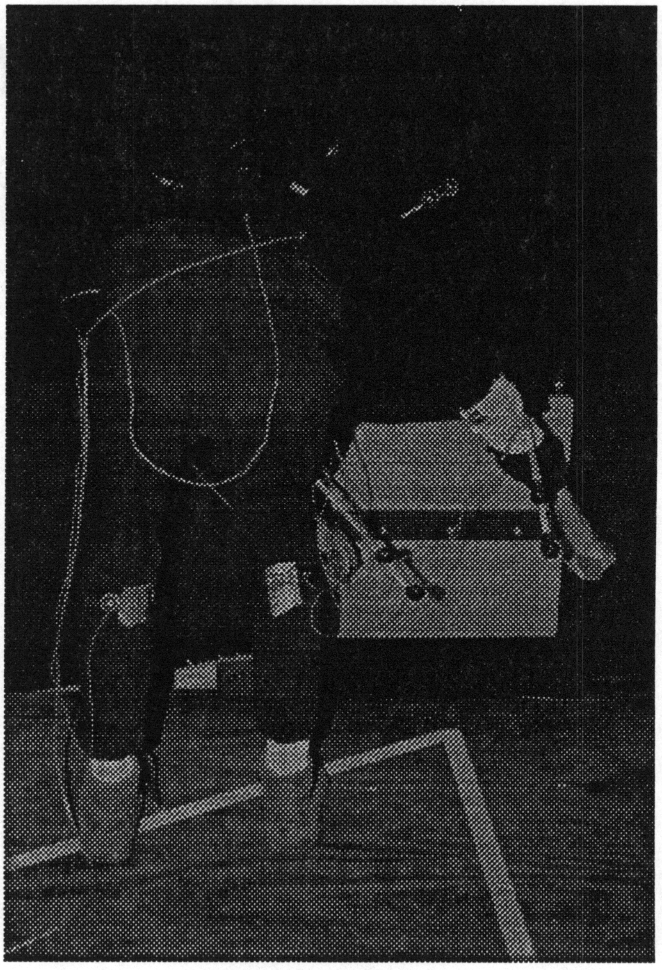

Figure 17. Treatment #6 - asymmetric lift, calf height, nonbelt condition.

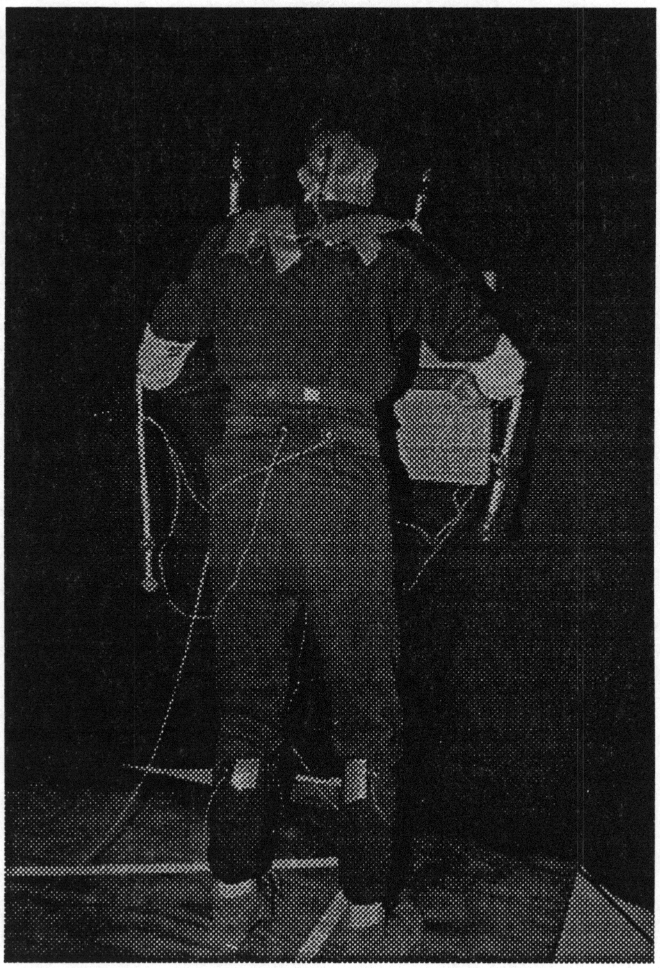

Figure 18. Treatment #7 - symmetric lift, standing elbow height, nonbelt condition.

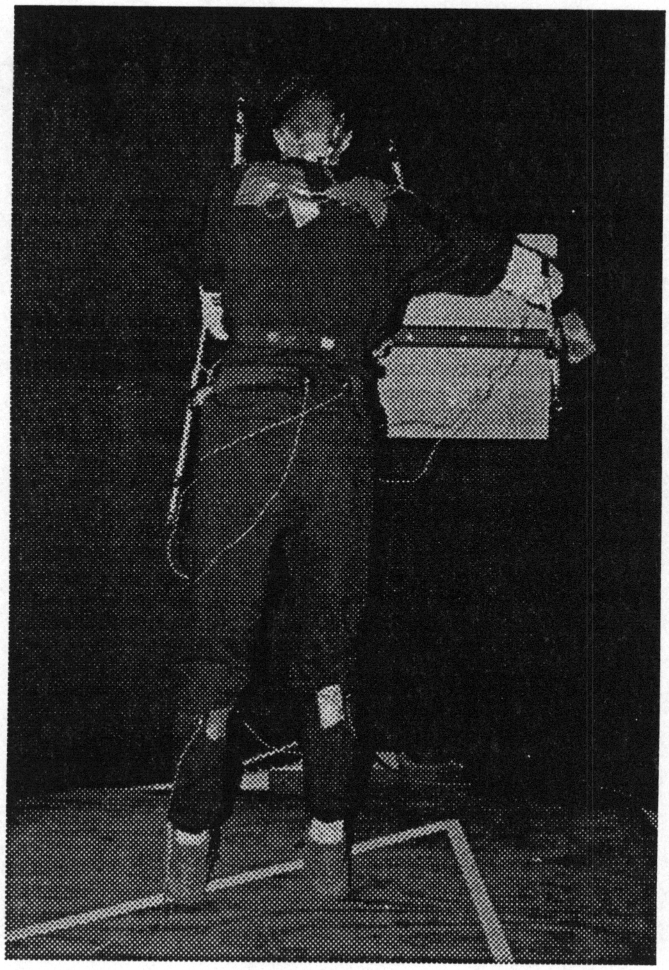

Figure 19. Treatment #8 - asymmetric lift, standing elbow height, nonbelt condition.

Upon arrival for the first session, the participant filled out an informed consent form and a physical fitness questionnaire (see Appendices A and B). If the participant was in good health and had no history of musculoskeletal injury or back pain , several anthropometric measurements (Table 13) were recorded.

The participant was then asked to practice each of the nonsupported trial conditions. During this time the experimenter marked, with colored tape, foot position and horizontal distances for the participant to follow during the experiment.

Upon arrival to the second session, each participant was asked to change into a set of black clothing, which was worn to reduce reflections. Each participant was also fitted with the abdominal belt and instructed on its proper use. The experimenter then attached infrared diodes of the WATSMART system to the locations discussed in section 3.3. Once the diodes were fixed, the participant was asked to approach the task area.

The participant was then asked to complete each of the 16 trials. Each participant was instructed to place the front of their feet on the appropriate tape markers. When in the correct location, the WATSMART system was connected to the IRED diodes. Once the participant grasped the lift handles of the apparatus, verbal cues were given as to when to begin the static lift.

For each condition, the participants performed a 6-second maximum voluntary exertion. Each exertion consisted of building up force for the first 2 seconds and then holding at maximum voluntary exertion for 4 seconds. Each participant was asked to exhale during the entire 6-second exertion. After the

Table 13. Recorded anthropometric measurements.

Measurement (units)	Description
Weight (kg)	
Stature (cm)	Vertical distance from floor to top of head, with shoes off and arms relaxed at the sides looking straight ahead.
Bicrestal breadth (cm)	The horizontal distance between the right and left ilia measured with a body caliper exerting sufficient pressure to compress the tissue overlying the bone.[1]
Trochanteric height (cm)	The vertical distance from the floor to the uppermost point on the trochanter of the femur.[1]
Vertebra prominens depth (cm)	The vertical distance from the top of the head to the vertebra prominens of the cervical region of the spine.
Standing elbow height (cm)	The vertical distance from the standing surface to the depression at the elbow between the humerus and the radius.
Calf height (cm)	The vertical distance from the standing surface to the maximum posterior protrusion of the gastrocnemius.

[1]As defined in Clauser, McConville, and Young (1969).

trial, the IRED diodes were disconnected from the WATSMART system and the participant was allowed to rest. Rest periods lasted for at least 3 minutes.

During the resting period, the experimenter checked the posture data collected for the previous trial to determine if the WATSMART system was able to "see" each diode. If more than a third of the postural data was missing between the 3rd and 5th second of the collection, the trial was redone. There are three reasons why the WATSMART system was not able to record a diode: 1) there was a short in a diode wire, 2) at least two cameras could not see every diode due to the type of posture assumed, or 3) clothing covered up a diode.

Upon the successful completion of the 16 different trials, each participant was paid, thanked for his/her time, and dismissed.

4. DATA ANALYSIS AND RESULTS

This chapter describes the data analysis and results after both postural and static strength data were collected. First, the methods used to reduce and analyze the collected data are discussed. The experimental results are then reported. Recall that the independent variables were the following: *Gender (G), Lift type (LT), Handle height (HH),* and *Belt condition (BC).*

4.1. Postural Data Analysis

The mean position of each IRED marker between the 3rd and 5th second for each collection was computed. This was accomplished through an averaging program, AVGPOS.EXE, written in Pascal. Hand position was computed through a similar program, HAVGPOS.EXE, also written in Pascal. Both averaging programs omitted missing data points, if present, from mean IRED position calculations. The averaged IRED marker locations were then transferred and run through several inter-linked Excel spreadsheets.

The first spreadsheet served two purposes. The first purpose was to determine the body landmark locations of those points which required two IRED markers (left elbow, left shoulder, vertebra prominens, right shoulder, and right elbow) to be identified during the posture data collection. The second purpose was to calculate the average location of each joint-center, relative to the room reference frame (Figure 20), over the two repetitions for like trial conditions. The average of each of the IRED marker locations was defined as:

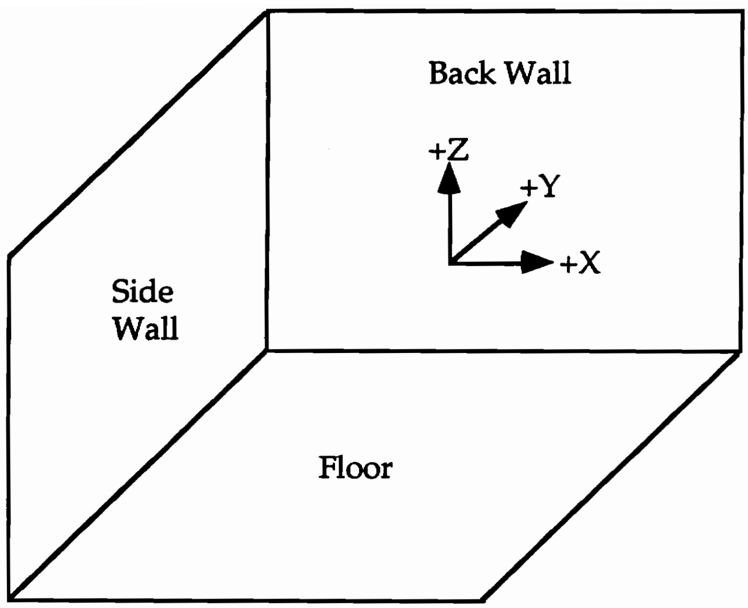

Figure 20. Orientation of the room coordinate system.

$$\frac{(a_{i1} + a_{i2})}{2} \qquad (4.1.),$$

where a_{i1} is the (x,y,z) coordinates of a given IRED marker location for the first repetition, a_{i2} is the (x,y,z) coordinates of a the same given IRED marker location for the second repetition, and i is the trial condition. This procedure was repeated for the eight different trial conditions for each participant.

A second Excel spreadsheet, linked to the first, was used to calculate the body reference frames for each condition. Using the *Gram-Schmidt orthogonalization process,* the Z-axis of the body reference frame for each condition was represented by the negative vector between the L3/L4 joint-center (defined as the body reference frame origin) and the midpoint between the calves of both legs.

Two planes were then defined orthogonal to the Z-axis established above. Two vectors orthogonal to the Z-axis, γ_2 and γ_3, were computed as:

$$\gamma_2 = \alpha_2 - \left[\frac{(\alpha_2|\gamma_1)}{(\gamma_1|\gamma_1)}\right]\gamma_1 \qquad (4.2),$$

$$\gamma_3 = \alpha_3 - \left[\frac{(\alpha_3|\gamma_1)}{(\gamma_1|\gamma_1)}\right]\gamma_1 - \left[\frac{(\alpha_3|\gamma_2)}{(\gamma_2|\gamma_2)}\right]\gamma_2 \qquad (4.3),$$

where γ_1 represented the Z-axis expressed as a vector. The variables α_2 and α_3 were defined as (1,0,0) and (0,1,0) respectively.

As part of the postural measurement protocol, a vector was measured from the L3/L4 joint-center to 11 centimeters lateral of that marker. This vector established to what degree the cutting plane had rotated about the midpoint of the calves of both knees. This point was not necessarily orthogonal to the Z-axis defined above (Figure 21). To account for this, the X-axis was defined by projecting the lateral vector defined above onto the orthoganal plane defined by γ_2 and γ_3. The X-axis was then expressed as:

$$\beta = \left[\frac{(\alpha|\gamma_2)}{(\gamma_2|\gamma_2)}\right]\gamma_2 + \left[\frac{(\alpha|\gamma_3)}{(\gamma_3|\gamma_3)}\right]\gamma_3 \qquad (4.4),$$

where α represented the lateral vector and γ_2 and γ_3 were as before.

The Y-axis was derived by calculating a right-hand vector orthogonal to both γ_1 (Z-axis) and β (X-axis). The Y-axis was expressed as:

$$\delta = \gamma_1 \times \beta \qquad (4.5).$$

Once all three axes were computed for each condition, the markers for that condition were translated into the body reference frame.

This second spreadsheet also computed, for each condition, three different angles representing the degree of torso extension/flexion, torso lateral bending, and axial twist of the shoulders. The degree of torso flexion/extension was calculated by drawing a vector connecting the vertebra prominens joint-center with the L3/L4 joint-center. This vector was then projected onto the YZ (sagittal) plane of the body reference frame. The Law of Cosines was then used to calculate the angle between the Y-axis and the torso alignment vector (Figure 23). A value of 90° indicates the absence of flexion

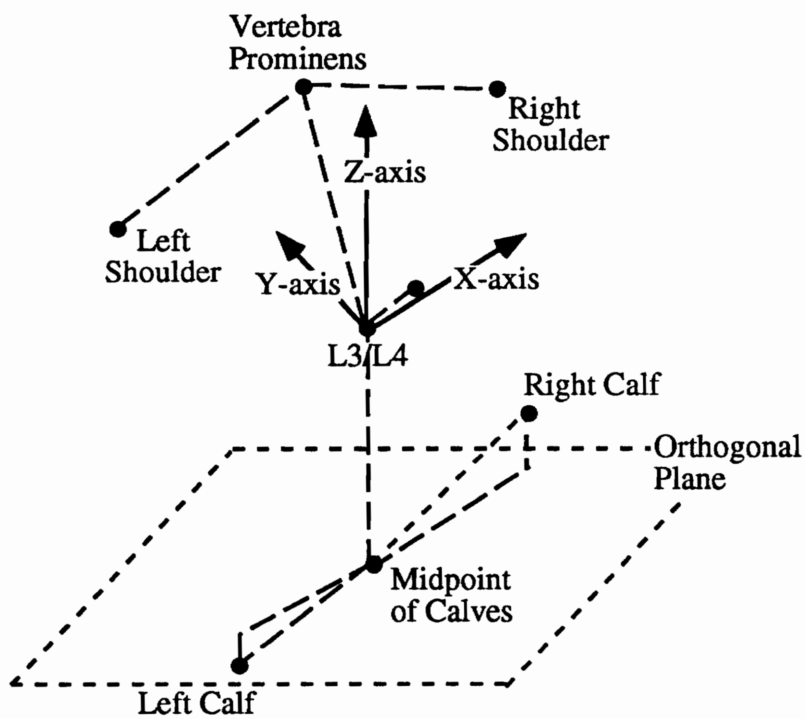

Figure 21. The axes of the body coordinate system as defined by the vectors β, δ, and γ_1 (X, Y, and Z - axes, respectively).

or extension. Values less than 90° indicated flexion while values greater than 90° indicated extension.

The calculation of the degree of torso lateral bending followed much the same procedure as described for the computation of the degree of torso flexion/extension angle. The difference lies in projecting the torso alignment vector onto the XZ (frontal) plane instead of the YZ plane. The Law of Cosines was then used to calculate the angle between the X-axis and the projected vector (Figure 23). A value of 90° indicated the absence of any lateral bending. Values less than 90° indicated lateral bending to the right (positive X) while values greater than 90° indicated lateral bending to the left.

To calculate the degree of axial twist of the shoulders in relation to the hips, a vector was drawn connecting the two shoulder joint-centers. This vector (shoulder vector) approximated the alignment of the shoulders. Another vector was drawn, connecting the L3/L4 joint-center with the lateral L3/L4 marker. This vector (hip vector) approximated the line in which the hips were located. Once these vectors were determined, they were both projected onto the XY (transverse) plane of the body reference frame. The Law of Cosines was then used to calculate the angle between the two vectors (Figure 24). This angle represented the degree of axial twist between the shoulders and the hips.

Torso Flexion/Extension Angle

The torso angles of flexion/extension were subjected to a repeated measures ANOVA. The ANOVA revealed that the main effect of *Handle*

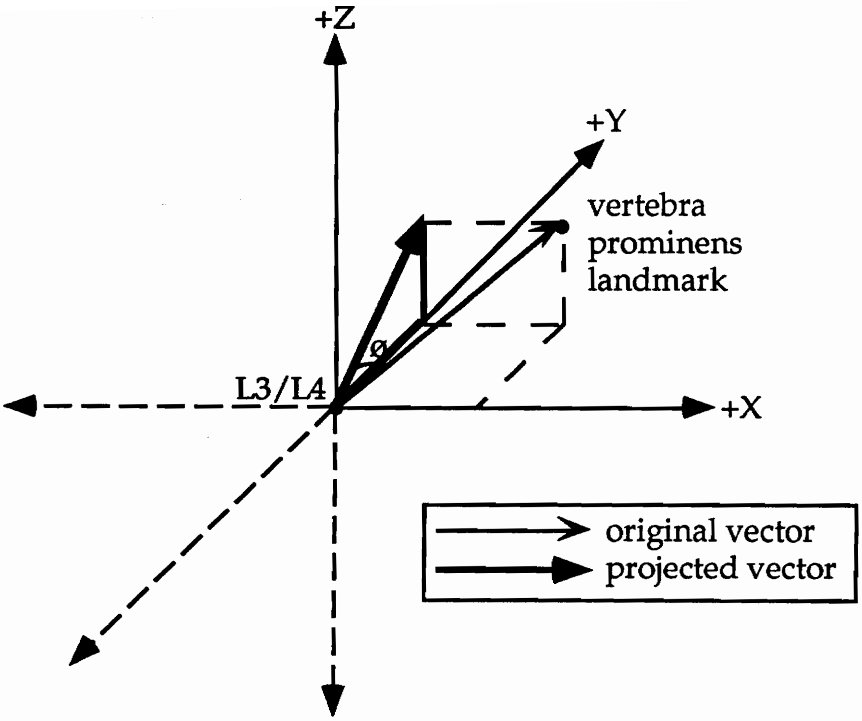

Figure 22. Torso extension/flexion angle as projected upon the YZ (sagittal) plane.

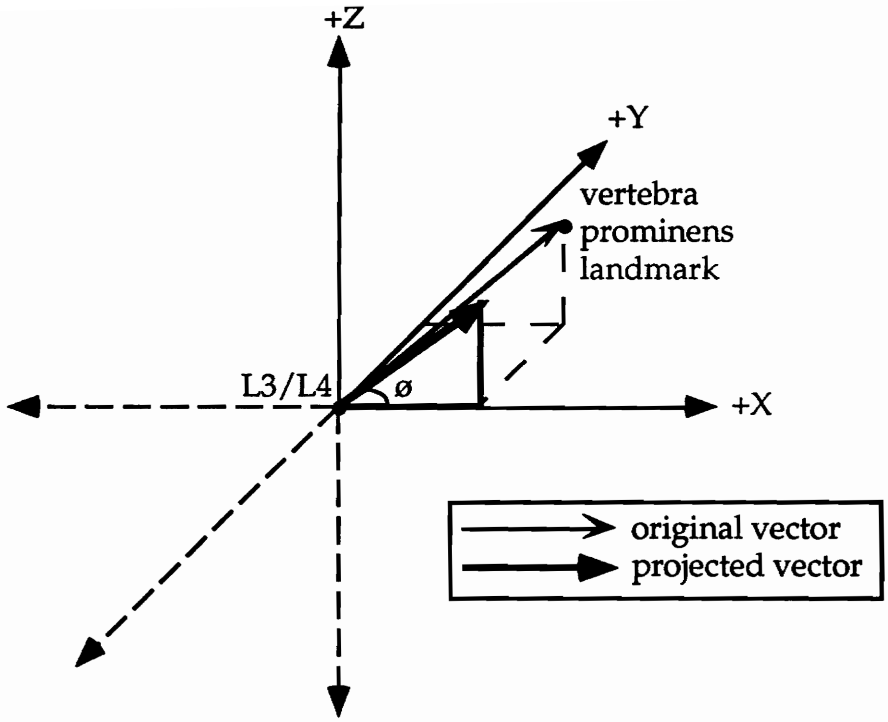

Figure 23. Torso lateral bending angle as projected upon the XZ (frontal) plane.

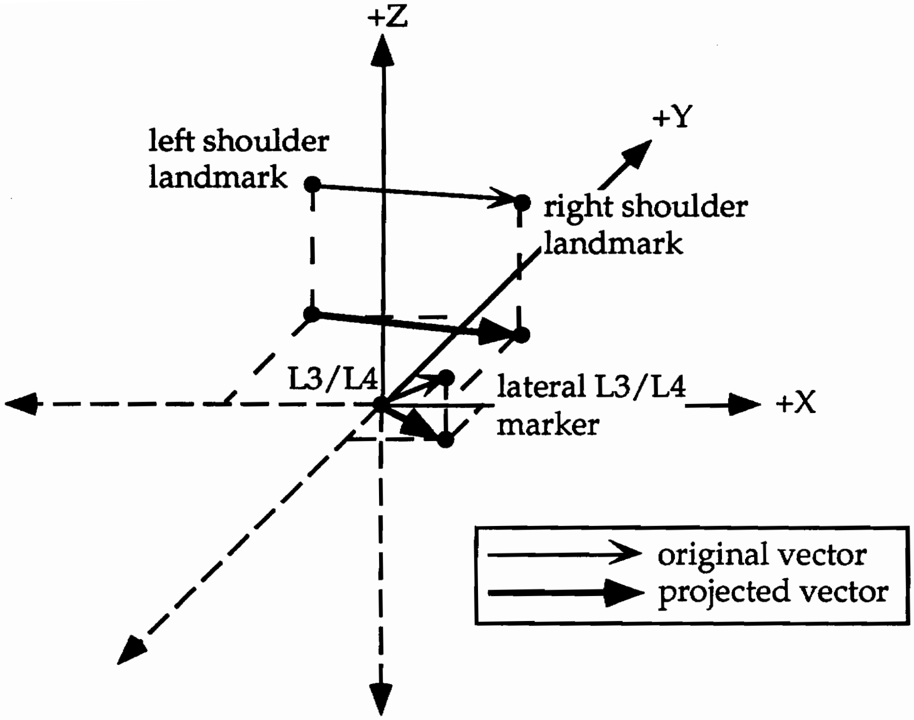

Figure 24. Axial twist angle as projected upon the XY (transverse) plane.

height and the three-way interaction of *Handle height* x *Lift type* x *Gender* were significant at $p<0.05$. Table 14 is the ANOVA summary table for the analysis. The significant main effect of *Handle height* ($p<0.001$) revealed that the torso was flexed more during the lower calf lift position (11.28°) compared to the standing elbow height (79.57°).

The three-way interaction of *Handle height* x *Lift type* x *Gender* is plotted in Figure 25. The angle of flexion stayed relatively constant for lifts initiated at the standing elbow height for any combination of Lift type and gender. For lifts initiated at the calf height, a significant difference in torso flexion angle was only observed between males (7.14°) and females (15.70°) during the performance of the asymmetric lift.

Torso Lateral Bending Angle

The torso angles of lateral bending were subjected to a repeated measures ANOVA. The ANOVA showed that the main effects of *Handle height* and *Lift type* and the two-way interaction of *Handle height* x *Lift type* were significant at $p<0.05$. Table 15 is the ANOVA summary table for the analysis. The significant main effect of *Handle height* ($p<0.001$) revealed that the torso was laterally bent to the right more during the lower calf lift position (51.93°) compared to the standing elbow height (86.54°). The significant main effect of *Lift type* ($p<0.001$) showed that the torso was laterally bent to the right more during the asymmetric lift condition (52.05°) compared to the symmetric lift condition (86.42°).

Table 14. ANOVA summary table for torso flexion/extension angle.

Source	df	Sum of Squares	Mean Square	F-Value	P-Value
G	1	568.389	568.389	3.906	.0682
Subject(Grp)	14	2037.245	145.517		
BC	1	4.955	4.955	.500	.4909
BC * G	1	.343	.343	.035	.8551
BC * Subject(Grp)	14	138.642	9.903		
HH	1	149261.659	149261.659	1958.355	.0001
HH * G	1	3.715	3.715	.049	.8285
HH * Subject(Grp)	14	1067.050	76.218		
LT	1	46.439	46.439	.693	.4192
LT * G	1	17.809	17.809	.266	.6143
LT * Subject(Grp)	14	938.494	67.035		
BC * HH	1	13.943	13.943	1.331	.2680
BC * HH * G	1	4.210	4.210	.402	.5364
BC * HH * Subject(Grp)	14	146.674	10.477		
BC * LT	1	2.380	2.380	.310	.5867
BC * LT * G	1	.285	.285	.037	.8502
BC * LT * Subject(Grp)	14	107.625	7.688		
HH * LT	1	71.194	71.194	1.537	.2354
HH * LT * G	1	337.925	337.925	7.295	.0172
HH * LT * Subject(Grp)	14	648.479	46.320		
BC * HH * LT	1	55.202	55.202	4.541	.0513
BC * HH * LT * G	1	1.993	1.993	.164	.6916
BC * HH * LT * Subject(Grp)	14	170.181	12.156		

Dependent: torso flexion/extension angle

Legend:

G: Gender (2 levels; Female, Male)
BC: Belt condition (2 levels; Belt, Nonbelt)
HH: Handle height (2 levels; Calf, Standing Elbow)
LT: Lift type (2 levels; Symmetric, Asymmetric)

Interaction Plot - Handle height x Lift type x Gender

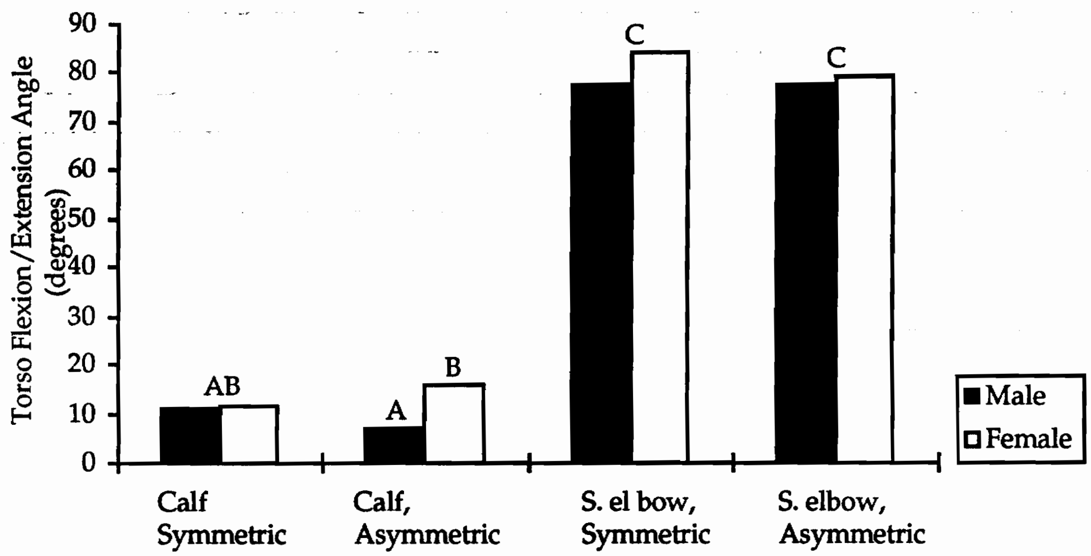

Figure 25. *Handle height* x *Lift type* x *Gender* interaction plotted against torso flexion/extension angle. (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

Table 15. ANOVA summary table for torso lateral bending angle.

Source	df	Sum of Squares	Mean Square	F-Value	P-Value
G	1	107.101	107.101	.337	.5706
Subject(Grp)	14	4445.551	317.539		
BC	1	61.729	61.729	.454	.5113
BC * G	1	311.342	311.342	2.291	.1524
BC * Subject(Grp)	14	1902.636	135.903		
HH	1	38328.810	38328.810	1.04E2	.0001
HH * G	1	3.761	3.761	.010	.9210
HH * Subject(Grp)	14	5168.853	369.204		
LT	1	37790.056	37790.056	1.38E2	.0001
LT * G	1	494.091	494.091	1.803	.2007
LT * Subject(Grp)	14	3836.869	274.062		
BC * HH	1	10.462	10.462	.073	.7911
BC * HH * G	1	249.580	249.580	1.739	.2084
BC * HH * Subject(Grp)	14	2008.799	143.486		
BC * LT	1	8.876	8.876	.057	.8152
BC * LT * G	1	161.000	161.000	1.029	.3276
BC * LT * Subject(Grp)	14	2190.481	156.463		
HH * LT	1	27318.976	27318.976	1.19E2	.0001
HH * LT * G	1	740.803	740.803	3.215	.0946
HH * LT * Subject(Grp)	14	3225.445	230.389		
BC * HH * LT	1	30.304	30.304	.204	.6585
BC * HH * LT * G	1	242.685	242.685	1.633	.2221
BC * HH * LT * Subject(Grp)	14	2080.402	148.600		

Dependent: torso lateral bending angle

Legend:

G: Gender (2 levels; Female, Male)
BC: Belt condition (2 levels; Belt, Nonbelt)
HH: Handle height (2 levels; Calf, Standing Elbow)
LT: Lift type (2 levels; Symmetric, Asymmetric)

The *Handle height* x *Lift type* ($p<0.001$) interaction is plotted in Figure 26. The angle of lateral bending of the torso stayed relatively unchanged for lifts initiated at the standing elbow height. In addition there was no difference in lateral bending angle between the lifts initiated at the calf and standing elbow heights for the symmetric lift condition. During the asymmetric lift condition, this difference became significant (63.83°).

Axial Twist of the Shoulders

The angles of axial twist of the shoulders in relation to the hips were subjected to a repeated measures analysis of variance (ANOVA). The ANOVA showed the three-way interaction of *Belt condition* x *Handle height* x *Lift type* was significant at $p<0.05$ ($p<0.04$). Table 16 is the ANOVA summary table for the analysis. No main effects were significant. Figure 27 shows the plot of this interaction. The calf asymmetric Belt condition was observed to be significantly different from the belt nonbelt calf asymmetric and belt nonbelt standing elbow height symmetric lift conditions.

4.2. Static Strength Data Analysis

Similar to the procedure followed for the posture data, the static strength data collected off the platform in A/D units was converted to newtons and the mean was computed between the 3rd and 5th second. MEANFORC.EXE, a program written in Pascal, handled this conversion. A repeated measures ANOVA was then performed upon the Y force component (upward lift force). Table 17 is the ANOVA summary table for the analysis.

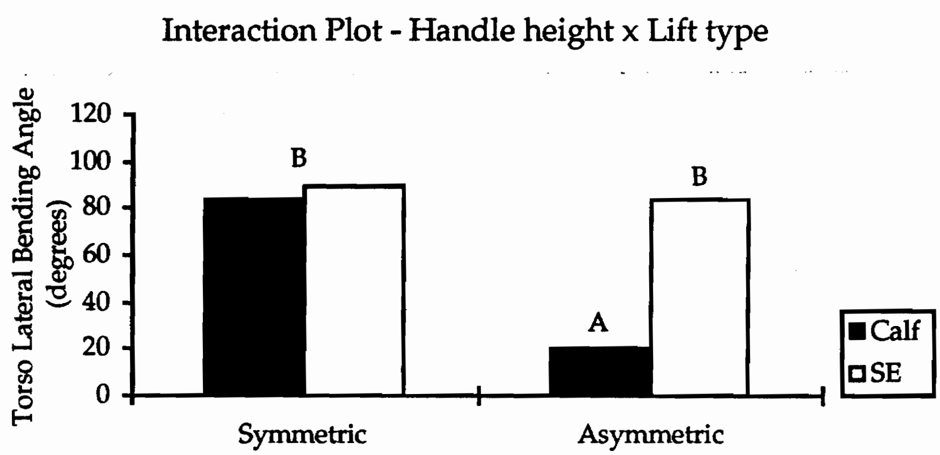

Figure 26. *Handle height* x *Lift type* interaction plotted against torso lateral bending angle. (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

Table 16. ANOVA summary table for angle of axial twist of the shoulders.

Source	df	Sum of Squares	Mean Square	F-Value	P-Value
G	1	2.165	2.165	.010	.9232
Subject(Grp)	14	3147.076	224.791		
BC	1	110.945	110.945	1.570	.2308
BC * G	1	127.102	127.102	1.798	.2013
BC * Subject(Grp)	14	989.596	70.685		
HH	1	4.663	4.663	.053	.8213
HH * G	1	28.448	28.448	.323	.5787
HH * Subject(Grp)	14	1232.544	88.039		
LT	1	66.442	66.442	3.002	.1051
LT * G	1	90.448	90.448	4.087	.0628
LT * Subject(Grp)	14	309.830	22.131		
BC * HH	1	2.632	2.632	.078	.7837
BC * HH * G	1	12.183	12.183	.362	.5569
BC * HH * Subject(Grp)	14	470.853	33.632		
BC * LT	1	4.834	4.834	.258	.6197
BC * LT * G	1	.241	.241	.013	.9114
BC * LT * Subject(Grp)	14	262.762	18.769		
HH * LT	1	41.391	41.391	.817	.3815
HH * LT * G	1	7.468	7.468	.147	.7069
HH * LT * Subject(Grp)	14	709.670	50.691		
BC * HH * LT	1	46.593	46.593	5.258	.0378
BC * HH * LT * G	1	37.889	37.889	4.276	.0577
BC * HH * LT * Subject(Grp)	14	124.055	8.861		

Dependent: axial twist angle

Legend:

G: Gender (2 levels; Female, Male)
BC: Belt condition (2 levels; belt, Nonbelt)
HH: Handle height (2 levels; Calf, Standing Elbow)
LT: Lift type (2 levels; Symmetric, Asymmetric)

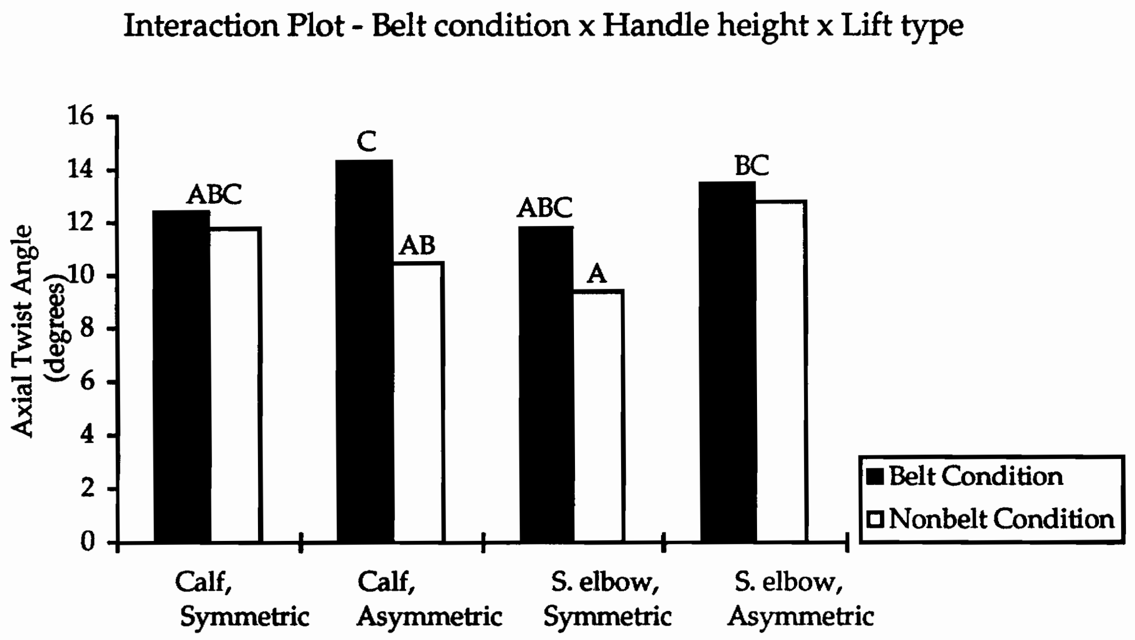

Figure 27. *Belt condition* x *Handle height* x *Lift type* interaction plotted against angle of axial twist of the shoulders. (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

Table 17. ANOVA summary table for upward lift force.

Source	df	Sum of Squares	Mean Square	F-Value	P-Value
G	1	753849.252	753849.252	17.178	.0010
Subject(Grp)	14	614385.011	43884.644		
BC	1	21.492	21.492	.048	.8291
BC * G	1	38.830	38.830	.087	.7719
BC * Subject(Grp)	14	6222.289	444.449		
HH	1	37551.701	37551.701	15.938	.0013
HH * G	1	1633.561	1633.561	.693	.4190
HH * Subject(Grp)	14	32985.754	2356.125		
LT	1	165316.813	165316.813	46.223	.0001
LT * G	1	19126.013	19126.013	5.348	.0365
LT * Subject(Grp)	14	50071.213	3576.515		
BC * HH	1	77.082	77.082	.221	.6454
BC * HH * G	1	3062.901	3062.901	8.789	.0102
BC * HH * Subject(Grp)	14	4878.733	348.481		
BC * LT	1	1752.394	1752.394	4.502	.0522
BC * LT * G	1	231.663	231.663	.595	.4533
BC * LT * Subject(Grp)	14	5449.952	389.282		
HH * LT	1	2881.733	2881.733	1.931	.1863
HH * LT * G	1	584.777	584.777	.392	.5414
HH * LT * Subject(Grp)	14	20889.734	1492.124		
BC * HH * LT	1	1113.035	1113.035	2.396	.1439
BC * HH * LT * G	1	.398	.398	.001	.9771
BC * HH * LT * Subject(Grp)	14	6502.952	464.497		

Dependent: static strength

Legend:

G: Gender (2 levels; Female, Male)
BC: Belt condition (2 levels; Belt, Nonbelt)
HH: Handle height (2 levels; Calf, Standing Elbow)
LT: Lift type (2 levels; Symmetric, Asymmetric)

The main effects of *Gender*, *Handle height*, and *Lift type*, along with the two-way interaction of *Lift type* x *Gender* and the three-way interaction of *Belt condition* x *Handle height* x *Gender* were found to be significant at $p<0.05$. The upward static lift strength generated by males (291.55 N) was greater than females (138.06 N). The values for lift strength with regards to the initial lift height revealed that the calf height resulted in a higher maximum voluntary contraction (231.93 N) as compared to the standing elbow height (197.68 N). It was also observed that symmetric lifts resulted in higher strength values (250.74 N) as compared to asymmetric lifts (178.87 N).

The *Lift type* x *Gender* ($p<0.04$) interaction is plotted in Figure 28. Each combination of Lift type and gender was significantly different from one another. In addition, the difference in strength generation for males with regard to Lift type was observed to be larger (96.33 N) than females (47.42 N).

The *Belt condition* x *Handle height* x *Gender* ($p<0.02$) interaction is plotted in Figure 29. Each combination of handle height and Belt condition for females was significantly different as compared to males. The nonbelt conditions for both males and females also revealed to be significantly different from one another.

4.3. Predicted L3/L4 Compression Data Analysis

The L3/L4 compressive force estimates were produced by a model using the double linear programming algorithm of Bean, Chaffin, and Shultz (1988)

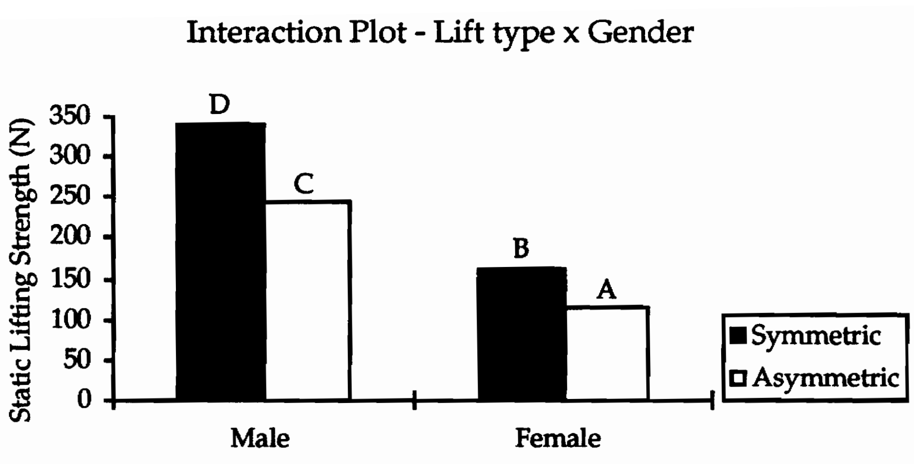

Figure 28. *Lift type* x *Gender* interaction plotted against upward lift force (N). (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

Interaction Plot - Belt condition x Handle height x Gender

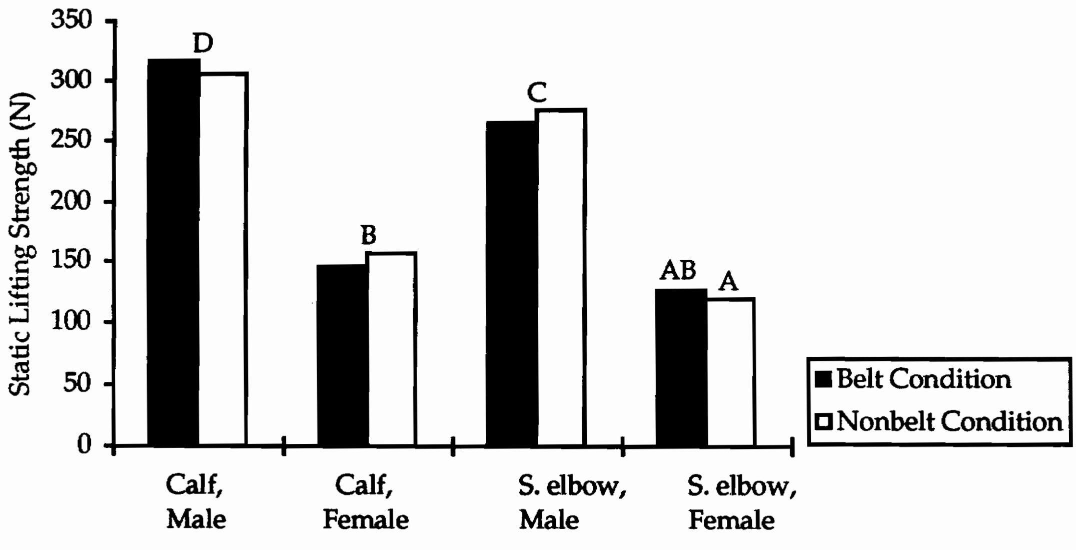

Figure 29. *Belt condition* x *Handle height* x *Gender* interaction plotted against upward lift force (N). (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

written in Mathematica. Inputs to the model along with the anthropometric measurements of weight, stature, vertebra prominens depth, trochanteric height, and bicrestal breadth included the joint-center locations in the room reference frame and the reaction forces at both hands. It was assumed the forces measured off the platform were divided equally among both hands. A repeated measures ANOVA was then performed upon the predicted compression at the L3/L4 intervertebral disc. Table 18 is the ANOVA summary table for the analysis.

The main effects of *Gender, Belt condition, Handle height*, and *Lift type*, along with the two-way interactions of *Handle height* x *Gender* and *Handle height* x *Lift type* were found to be significant at $p<0.05$. L3/L4 compression values for males (3659.52 N) were significantly larger than for females (2165.82 N). Using a belt resulted in a lower compression value (2737.87 N) than when a belt was not used (3087.47 N). Lifts initiated at the calf height resulted in higher compression values (3718.04 N) and those lifts initiated at the standing elbow height (2107.30 N). It was also revealed that symmetric lifts resulted in higher compression values (3094.48 N) than asymmetric lifts (2730.86 N).

The *Handle height* x *Gender* ($p<0.05$) interaction is plotted in Figure 30. It was observed that the male-standing elbow height condition and the female-calf height condition were not significantly different. Males on a whole were subjected to higher compression values than females which supports the significant *Gender* effect.

The *Handle height* x *Lift type* ($p<0.01$) interaction is plotted in Figure 31. The standing elbow height was unchanged across Lift type. In addition, the difference in observed compression between calf and standing elbow height

Table 18. ANOVA summary table for L3/L4 intervertebral disc spinal compression.

Source	df	Sum of Squares	Mean Square	F-Value	P-Value
G	1	71396181.796	7.14E7	20.542	.0005
Subject(Grp)	14	48659142.323	3475653.023		
BC	1	3911030.786	3911030.786	12.354	.0034
BC * G	1	477.363	477.363	.002	.9696
BC * Subject(Grp)	14	4432100.024	316578.573		
HH	1	83024042.158	8.302E7	56.985	.0001
HH * G	1	6847337.212	6847337.212	4.700	.0479
HH * Subject(Grp)	14	20397159.868	1456939.991		
LT	1	4231092.865	4231092.865	9.392	.0084
LT * G	1	38439.361	38439.361	.085	.7745
LT * Subject(Grp)	14	6307080.356	450505.740		
BC * HH	1	37829.655	37829.655	.079	.7824
BC * HH * G	1	306643.264	306643.264	.643	.4361
BC * HH * Subject(Grp)	14	6677696.651	476978.332		
BC * LT	1	317731.335	317731.335	.732	.4065
BC * LT * G	1	291654.835	291654.835	.672	.4260
BC * LT * Subject(Grp)	14	6074252.024	433875.145		
HH * LT	1	7618632.216	7618632.216	10.077	.0068
HH * LT * G	1	970450.702	970450.702	1.284	.2763
HH * LT * Subject(Grp)	14	10584159.138	756011.367		
BC * HH * LT	1	470409.625	470409.625	1.192	.2934
BC * HH * LT * G	1	9460.637	9460.637	.024	.8792
BC * HH * LT * Subject(Grp)	14	5524958.590	394639.899		

Dependent: L3/L4 compression

Legend:

G: Gender (2 levels; Female, Male)
BC: Belt condition (2 levels; Belt, Nonbelt)
HH: Handle height (2 levels; Calf, Standing Elbow)
LT: Lift type (2 levels; Symmetric, Asymmetric)

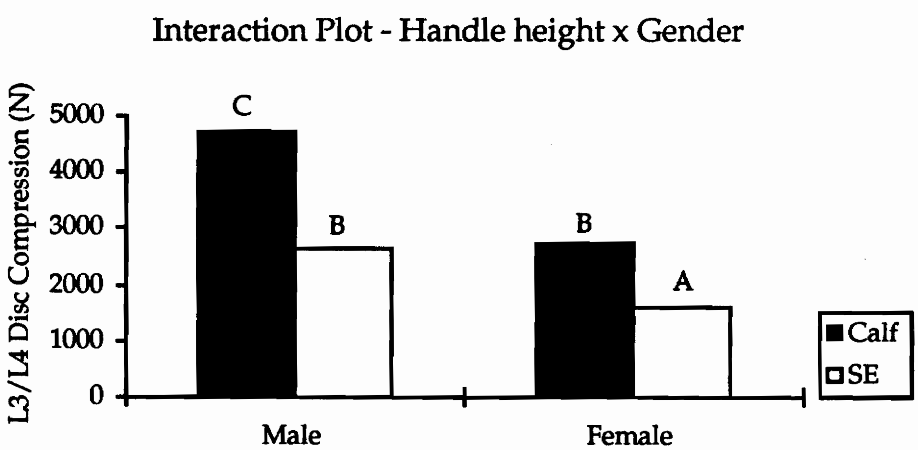

Figure 30. *Handle height* x *Gender* interaction plotted against L3/L4 intervertebral disc spinal compression (N). (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

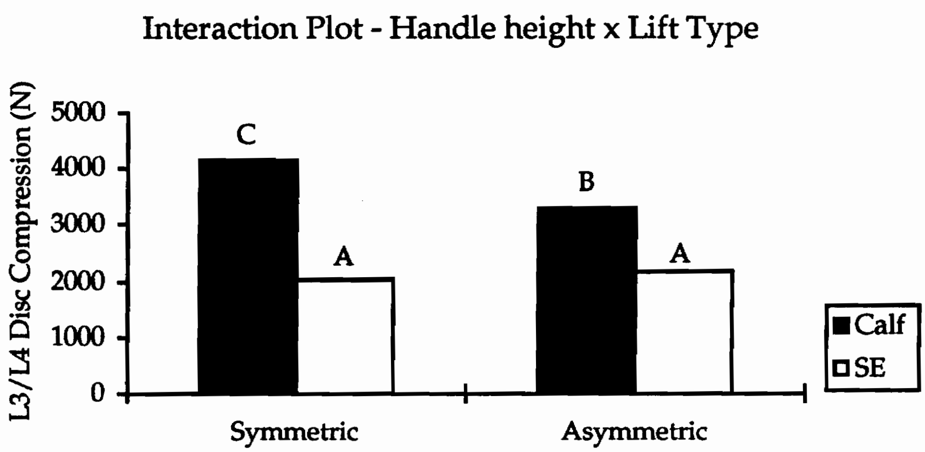

Figure 31. *Handle height* x *Lift type* interaction plotted against L3/L4 intervertebral disc spinal compression (N). (Conditions with same letters are not significantly different using a Newman-Keuls post hoc comparison test.)

significantly decreased across the Lift type conditions. (2098.68 N for symmetric verses 1122.81 N for asymmetric). This finding is consistent with the significant result for the main effect of *Lift type*.

5. DISCUSSION

Recall the three objectives for this thesis outlined in Chapter 1:

Objective 1: To develop a procedure and measure the difference in torso posture across static lift trials differing in symmetry and height across the belt and nonbelt conditions of a commercially available abdominal support belt.

Objective 2: To measure and compare the static lift strength across the belt and nonbelt conditions.

Objective 3: To predict and compare the levels of low-back compressive force at the L3/L4 vertebral disc across both belt and nonbelt use conditions using a three-dimensional biomechanical model.

This chapter discusses the results obtained from the data analysis with regards to the above stated objectives. Areas of future research are also identified.

5.1. Torso Posture Change

The first objective was to determine if the users of a commercially available abdominal support belt assumed different torso postures when the belt was worn as compared to when it was not worn. As revealed in the results section, torso posture, with the exception of axial twist, was significantly effected only by the type of lift and the height of the static lift. These significant differences were expected due to the fact that the experimental conditions required the assuming of different postures in order to perform the various static lifts. It was also observed that females did not flex as much as males when calf asymmetric lifts were performed. It is hypothesized that differences in anthropometry between each gender, especially with regard to upper torso structures and low torso differences (i.e. differences in gait, larger pelvic structures for females, etc.), are responsible for the significant difference.

Axial twist was significantly higher when the belt was used compared to when the belt was not used for the calf height asymmetric lift condition. This result may have been caused by the belt not allowing the hips to turn with the upper torso when low static lifts were performed. To compensate for the lack of twisting of the lower torso region, the twist of the upper torso became more pronounced. This condition may actually increase the risk of injury to the back due to the increased twist of the spine over a smaller region.

5.2. Static Strength Generation

The second objective addressed whether the use of an abdominal support belt increased static lift strength. Differences in lift strength values were not observed across the trial conditions with regard to support belt use. These findings agree with Woodhouse et al. (1990) in that wearing a belt does not improve lift capacity.

As expected gender, handle height, and Lift type were found significant. Higher lift strength values were observed at the lower symmetric condition due to the fact the large muscles of the leg could be used to apply force more "in-line" with the vertical lift. The standing elbow height required participants to lift more with their arms rather than their larger leg muscles. Asymmetric lift strength was observed to be lower than lifts performed in the sagittal plane. An explanation for this, as reported by Garg (1983), is that the upward force application also included a larger lateral force in an attempt to move the load closer to the body's center of gravity as compared to symmetric lifts. Related to

Garg's explanation, reduced lift force may also be due to each of the participants attempting to maintain their balance during the exertion.

5.3. L3/L4 Compression

The final objective of this thesis was to determine if predicted L3/L4 spinal compression is affected by the use of a support belt. The results indicated all of the experimental variables were significant. A larger value for the nonbelt condition was observed as compared to the Belt condition. This result was surprising due to the fact that posture and static strength generation for the Belt condition was found not significant.

One explanation for a significant belt result for compression may be due to the use of the belt affecting posture other than that of the torso. Changes in the distance that participants stood from the lift handles, especially for the standing elbow height condition, may not have become apparent through the analysis of torso posture.

Another explanation deals with the assumption that equal reaction force was experienced at both hands. For asymmetric exertions, there may have been a higher lateral component for one hand compared to the other due to the nature of the posture required to reach both lift handles.

A third explanation deals with the sensitivity of the biomechanical model with regard to the establishment of the Z-axis of the body reference frame. Dysart et al. (1993) had investigated the sensitivity of the Bean et al. (1988) model and discovered that errors in deriving the cutting plane for the body reference plane may have occurred due to errors in positioning of the shoulder or torso markers other changes in posture than that of the torso.

Those errors and changes may have directly lead to incorrect rotation of the Z-axis of the body reference plane. With regard to the research performed by Dysart et al. (1993), changes in compression due to rotation of the Z-axis of the body reference plane alone resulted in a maximum increase compression and maximum decrease of compression of 227.93 N and -148.48 N, respectively. Other factors influencing sensitivity are related to the inherent weaknesses of the Bean et. al (1988) model in relation to asymmetric lift conditions. As discussed in section 2.6., this model does not predict differential levels of erector spinae muscle force during lateral bending motion and does not take into account antagonistic muscle activity.

6. CONCLUSION

This research investigated the effects of the use of the Decade support belt on torso posture and lifting strength generation. Results show angle of axial twist for low asymmetric static exertions increased. Torso flexion/extension angle, torso lateral bending angle, and static strength generation were unaffected by belt use. Predicted compression at the L3/L4 intervertebral disc was significantly lower for the belt condition compared to unsupported condition. However, compression predictions are suspect due to limitations of the prediction model with regard to lateral loading of the trunk.

It is very tempting, although inappropriate, to generalize these results to other belt designs due to the fact that the body of research regarding commercially available support belts is very limited. Much of the research to date has focused on supports which were designed for immobilization of the spine, not for the performance of manual materials handling activities. Outlined below are future avenues for research involving commercial belt designs.

- ***Investigate the physical effect of other commercial belt designs.***

As discussed in Section 2.8., a variety of different belt designs made with different materials are available on the commercial market. It would be inappropriate to generalize the findings of this research to the other available designs. Similar testing on a sampling of the other available designs would add validity to generalizations made about the effectiveness or ineffectiveness of commercial abdominal supports.

- ***Investigate the physical effects of belt use on submaximal exertions.***

This research centered around a task that required participants to maximally exert upon the lift apparatus. Most lifts in industry do not require maximal exertions. Future testing should address this issue.

- ***Examine the effect of anthropometric differences of the wearers with regard to support belt use.***

The human comes in many sizes and shapes. It is important to understand how belt use affects those who, for example, are obese or very thin.

- ***Examine physical belt effects upon back school training of proper lifting techniques.***

The prescription of back supports to workers usually is accompanied by some type of training (whether classroom-oriented or other) on proper lifting techniques. Support belt users are instructed to follow good lifting techniques in addition to the wearing of the belt during the performance of lifting tasks. For this research, participants were instructed to assume whatever posture was most comfortable. However, these postures may or may not have been considered "proper". Future research should investigate how the assuming of proper postures if affected by commercial support belt use.

- ***Examine the physical effect of improper wearing of support belts.***

Up to this point, research has focused on the investigation into belt effects assuming that the supports are worn properly. It is also important to determine the physical effect of support belts when they are improperly worn. It is unrealistic to assume that workers are properly wearing and following the instructions provided with each belt.

- ***Examine the physical effect of prolonged use upon abdominal strength.***

It has been suggested that, as a result of prolonged use of support belts, a reduction of abdominal strength occurs. With this reduction, the incidence of injury is believed to increase when heavy lifts are then performed when a belt is not worn. Noncompliance situations are very possible and sometimes unavoidable. For example most who are involved in manual materials handling at work and use support belts usually do not use support belts for the same tasks around the home.

- ***Implement dynamic testing of the physical effect of support belts on lifting tasks.***

Manual materials handling tasks are all dynamic in nature. Using a static task to the examine the physical effect of support belt use, although effective, may not show other higher level interactions that can be revealed as a result of dynamic motion analysis.

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APPENDIX A: Informed Consent Form

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants of Investigative Projects

I. THE PURPOSE OF THIS RESEARCH/PROJECT

You are invited to participate in a study investigating the effect of abdominal support belt use upon lifting. Further details about the project will be explained at the end of the experiment.

II. PROCEDURES

You will be asked to perform different maximal static lifting trials while wearing and not wearing an abdominal support belt. All of the experimental tasks will involve *maximal safe exertions.* It is important when you complete these trials that you exert what you feel to be your maximum safe effort, that is the maximum force you can exert without risking injury. Only you can be the judge of what your maximum safe effort is, and you should not exceed that level, nor will anyone associated with this experiment ask you to do so.

The time and conditions required for you to participate in this project is approximately 2.5 hours over two days. You will be asked to wear both black sweat pants and a black sweat shirt which will be provided. Infrared emitting diodes (IREDS) will then be attached to your elbows, shoulders, and neck, and low-back. Then you will be asked to complete approximately 16 lifting tasks. A minimum of two minutes of rest will be allowed between trials.

The possible risks or discomfort to you as a participant may be some muscle fatigue and soreness after the completion of the experiment; however, this fatigue should be short-lived and pose no further complication or discomfort to you. There is no risk of electric shock due to the IREDS or any other part of the testing apparatus.

Safeguards that will be used to minimize your risk or discomfort are screening of your health history, close monitoring by the experimenter, short duration static lifts, and rest breaks between each trial.

III. BENEFITS OF THIS PROJECT

Your participation in this project will provide information regarding the physical effect of wearing abdominal support belts upon the user during lifting. This type of information can provide guidance to safety personnel upon the effectiveness of abdominal support belts as personal protective equipment during manual materials handling activities in industry.

No guarantee of benefits has been made to encourage you to participate.

You may receive a synopsis or summary of this research when it is completed. Please leave (or bring back) a self-addressed envelope.

IV. EXTENT OF ANONYMITY AND CONFIDENTIALITY

The results of this study will be kept strictly confidential. At no time will the researchers release the results of the study to anyone other than individuals working on the project without your written consent. The information you provide will have your name removed and only a subject number will identify you during analyses and any written reports of this research.

This experiment will be video taped. These tapes will only be reviewed by Brian R. Sherman and will be erased after this research has been completed.

V. COMPENSATION

Monetary

For participation in the project you will receive $5 for each hour completed.

VI. FREEDOM TO WITHDRAW

You are free to withdraw from this study at any time without penalty. If you chose to withdraw, you will be compensated for the portion of the time of the study in which you participated.

There may be circumstances under which the investigator may determine that you should not continue as a participant of this project. These might include fatigue or discomfort. If this occurs, you will be compensated for the portion of the project completed.

VII. APPROVAL OF RESEARCH

This research project has been approved, as required, by the Institutional Review Board for projects involving human subjects at Virginia Polytechnic Institute and State University, by the Department of Industrial and Systems Engineering.

VIII. PARTICIPANT'S RESPONSIBILITIES

I know of no reason I cannot participate in this study. I have the responsibility to complete the physical fitness questionnaire.

____

Signature

IX. PARTICIPANT'S PERMISSION (tear off at dashed line and give to subject)

I have read and understand the informed consent and conditions of this project. I have had all my questions answered. I hereby acknowledge the above and give my voluntary consent for participation in this project.

If I participate, I may withdraw at any time without penalty. I agree to abide by the rules of this project.

Should I have any questions about this research or its conduct, I will contact:

Brian R. Sherman, Investigator	231-6053
Dr. Jeffrey Woldstad, Faculty Advisor	231-4927
Chair Institutional Review Board, Research Division	231-9359

APPENDIX B: Participant Physical Fitness Questionnaire

PARTICIPANT PHYSICAL FITNESS QUESTIONNAIRE

Subject's Name:____
SSN:_________________________
Address:____
Telephone Number:(______)________________
Gender:______________ Date of Birth:________________________

Which best describes your present physical condition (circle one):

Poor Fair Good Excellent

Please describe any physical activities you presently participate in on a regular basis:

________________________________ : __________times per week
________________________________ : __________times per week

Please answer Yes or No to the following questions.
Have you ever had a hernia?________

Have you ever had a back injury?__________

Have you had any noticeable back pain during the last year?__________

Have you ever had any joint dislocations, broken bones, or other physical injuries in the last year?_______

Have you ever had any other serious musculoskeletal injury?________

Are you presently taking any medication or drugs?_______
If yes, please explain:____
____

Do you presently have any physical impairment or injury worth noting?____
If yes, please explain:____
____

Can you think of any injury, or illness you might have which could be aggravated by physical activity or participation in this experiment?____
If yes, please explain:____
____

____
Signature and Date

APPENDIX C: Biomechanical Model

BIOMECHANICAL MODEL

This appendix section describes the three-dimensional, static biomechanical model used to estimate the low-back compressive forces. The discussion is divided into three main parts. The first part describes the model component used to estimate the externally caused moment about the L3/L4 joint. This moment was initially expressed in a coordinate system orthogonal to the walls and floor of the laboratory. The second part outlines the procedures used to transform the moments and forces computed in the room reference frame to the reference frame defined by the body. The third part describes the Bean, Chaffin, and Schultz (1988) double linear programming algorithm which was used to estimate the internal trunk muscle forces required to compensate for the external moment. This part also describes how the internal and external forces combined to estimate the compressive force on the L3/L4 joint.

Moments and Forces due to the External Load

The external moment at the L3/L4 level was estimated by performing a static free body analysis on the links of the body above L3/L4. Recall that a static biomechanical system is governed by the equations of static equilibrium and in this case, can be expressed as:

$$\begin{aligned} \sum \mathbf{F}_{\text{external}} + \sum \mathbf{F}_{\text{internal}} &= 0 \\ \sum \mathbf{M}_{\text{external}} + \sum \mathbf{M}_{\text{internal}} &= 0 \end{aligned} \qquad (1.1),$$

where **F** and **M** are three-dimensional vectors representing the internal and external forces and moments acting on the body.

Data needed for the analysis are hand force vectors, posture coordinates, and link centers-of-mass and weights. The posture coordinates represent the positions of joint end points and centers-of-rotation in a three-dimensional coordinate system whose origin was located at the L3/L4 joint. The coordinate system is orthogonal to the walls and floor of the laboratory and is shown in Figure 1. The force vectors and position coordinates are both measured in this reference frame. This reference frame is referred to in the text as the *room reference frame*. The variables used in the free body analysis are listed in Table 1. Center-of-mass locations for the limbs are calculated based on a percentage of link length reported by Webb Associates (1978). The torso center-of-mass formula was taken from Clauser, McConville, and Young (1969). Individual segment weights are computed based on percentages of total body weight as described in Webb Associates (1978) and in

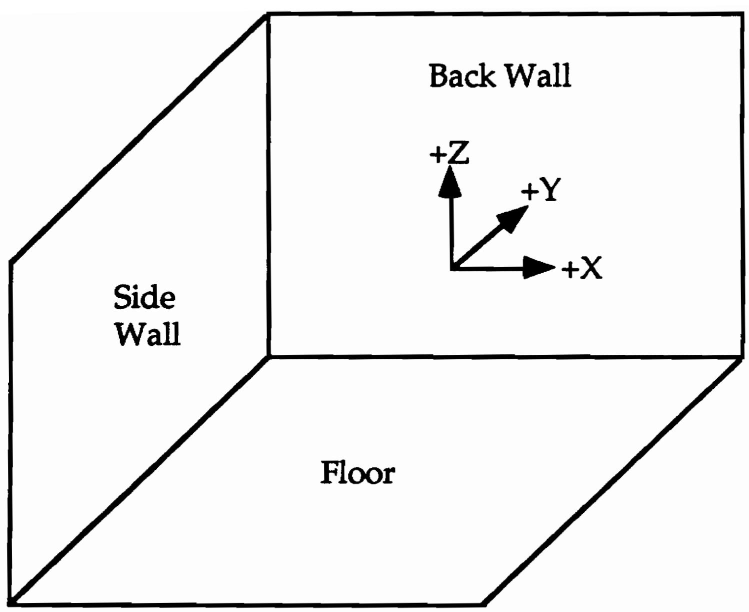

Figure 1. Orientation of the room coordinate system.

Table 1. Variables Used in External Force Model. Moment Arms r^j (cm) and Forces **F** (N)

Variable	Description	Calculation
r^j_{LA}	lower arm center-of-rotation (elbow) to hand grip location for side j	measured during experiment
$r^j_{LA\text{-}COM}$	lower arm center-of-rotation to lower arm center-of-mass for side j	$0.43(r^j_{LA})$
r^j_{UA}	upper arm center-of-rotation (shoulder) to lower arm center-of-rotation for side j	measured during experiment
$r^j_{UA\text{-}COM}$	upper arm center-of-rotation (shoulder) to upper arm center-of-mass for side j	$0.436\ (r^j_{UA})$
r^j_{Torso}	torso center-of-rotation (L3/L4) to upper arm center-of-rotation for side j	measured during experiment
$r_{Torso\text{-}COM}$	torso center-of-rotation to torso center-of-mass	(head to trunk length) - (.621*bicrestal breadth + .582* head to trunk length - .181*stature + 14.050)[1]
F^j_{Hand}	external reaction force acting on hand for side j	measured during experiment
$F_{LA\ Wt}$	force due to weight of lower arm (same for both sides)	0.023(total body weight)[2]
$F_{UA\ Wt}$	force due to weight of upper arm (same for both sides)	0.028(total body weight)[2]
$F_{Torso\ Wt}$	force due to weight of torso (includes head and neck)	0.58(total body weight)

[1] Refer to Clauser, McConville, and Young (1969) for explanation of terms.
[2] The weight of the total arm is divided into upper arm and lower arm (forearm and hands) segments according to the percentages reported in Webb Associates (1978).

Clauser, McConville, and Young (1969). The calculation of the externally caused moment about L3/L4 was then a process of moment addition. The model first calculates the three-dimensional vector moment at the elbow for each side of the body, which is expressed as:

$$\mathbf{M}^j_{\text{elbow}} = \mathbf{M}^j_{\text{ext force}} + \mathbf{M}^j_{\text{seg wt}} \tag{1.2}$$

where $\mathbf{M}^j_{\text{ext force}}$ is the moment about the elbow due to the hand force acting at the hand grip location for side j. The component $\mathbf{M}^j_{\text{segwt}}$ is the moment due to the weight of the lower arm rotating about the elbow for side j. In expanded form, the equation can be expressed as:

$$\mathbf{M}^j_{\text{elbow}} = (\mathbf{r}^j_{\text{LA}} \times \mathbf{F}^j_{\text{Hand}}) + (\mathbf{r}^j_{\text{LA-COM}} \times \mathbf{F}_{\text{LA Wt}}) \tag{1.3}$$

The model next calculates the moment about each shoulder, which is expressed as:

$$\mathbf{M}^j_{\text{shoulder}} = \mathbf{M}^j_{\text{ext force}} + \mathbf{M}^j_{\text{seg wt}} + \mathbf{M}^j_{\text{elbow}} \tag{1.4}$$

where $\mathbf{M}^j_{\text{ext force}}$ is now the sum of the hand force and lower arm weight acting at the elbow (end of the upper arm link) for side j. The vector $\mathbf{M}^j_{\text{seg wt}}$ now represents the moment due to the weight of the upper arm rotating about the shoulder for side j. The quantity $\mathbf{M}^j_{\text{elbow}}$ is the same quantity from Equation 2.3. In expanded form, the moment about each shoulder is:

$$\mathbf{M}^j_{\text{shoulder}} = \left[\mathbf{r}^j_{\text{UA}} \times (\mathbf{F}^j_{\text{Hand}} + \mathbf{F}_{\text{LA Wt}})\right] + (\mathbf{r}^j_{\text{UA-COM}} \times \mathbf{F}_{\text{UA Wt}}) + \mathbf{M}^j_{\text{elbow}} \tag{1.5}$$

The model finally calculates the moment about the L3/L4 joint, expressed as:

$$\mathbf{M}_{\text{L3/L4}} = \sum_{j=1}^{2} \mathbf{M}^j_{\text{ext force}} + \mathbf{M}_{\text{seg wt}} + \sum_{j=1}^{2} \mathbf{M}^j_{\text{shoulder}} \tag{1.6}$$

where $\mathbf{M}^j_{\text{ext force}}$ represents the moment due to hand force, lower arm weight, and upper arm weight acting about the shoulder rotating about the L3/L4 joint. The vector $\mathbf{M}_{\text{seg wt}}$ depicts the moment due to the weight of the torso (including head and neck) rotating about the L3/L4 joint. The moments due to external forces and the moments about the shoulder must be summed across sides because the L3/L4 moment is that of a singular joint. In expanded form, the moment about L3/L4 is:

$$\mathbf{M}_{L3/L4} = \sum_{j=1}^{2}\left[\mathbf{r}^{j}_{\text{Torso}} \times (\mathbf{F}^{j}_{\text{Hand}} + \mathbf{F}^{j}_{\text{LA Wt}} + \mathbf{F}^{j}_{\text{UA Wt}})\right]$$
$$+ (\mathbf{r}_{\text{Torso-COM}} \times \mathbf{F}_{\text{Torso Wt}}) + \sum_{j=1}^{2} \mathbf{M}^{j}_{\text{shoulder}} \qquad (1.7).$$

The model also resolves the external forces acting on the body, as these must be added when spinal disc compressive force is calculated. The external forces acting at L3/L4 are expressed as:

$$\mathbf{F}_{L3/L4} = \mathbf{F}_{\text{ext force}} + \mathbf{F}_{\text{seg wt}} \qquad (1.8),$$

where $\mathbf{F}_{\text{ext force}}$ represents the forces due to external loads and $\mathbf{F}_{\text{seg wt}}$ represents the total force due to gravity acting upon individual body segments above L3/L4. The equation in expanded form is:

$$\mathbf{F}_{L3/L4} = \sum_{j=1}^{2} \mathbf{F}^{j}_{\text{Hand}} + 2\mathbf{F}_{\text{LA Wt}} + 2\mathbf{F}_{\text{UA Wt}} + \mathbf{F}_{\text{Torso Wt}} \qquad (1.9).$$

To this point, all positions and moments have been expressed relative to the room reference frame. The forces and moments now must be transformed and expressed in the reference frame of the body.

Transformation to Body Reference Frame

The optimization-based model used to estimate low-back muscle forces requires that the resultant forces and moments acting at the L3/L4 joint be expressed in a coordinate system defined with respect to position of the body as opposed to the laboratory. This *body reference frame* was defined by a cutting plane orthogonal to the torsional axis of the L3/L4 vertebral section. Figure 2 shows the orientation of this coordinate system. A geometric model was developed to calculate a transformation matrix for converting forces and moments in the room reference frame to the body reference frame. A description of the transformation process used in the biomechanical model follows.

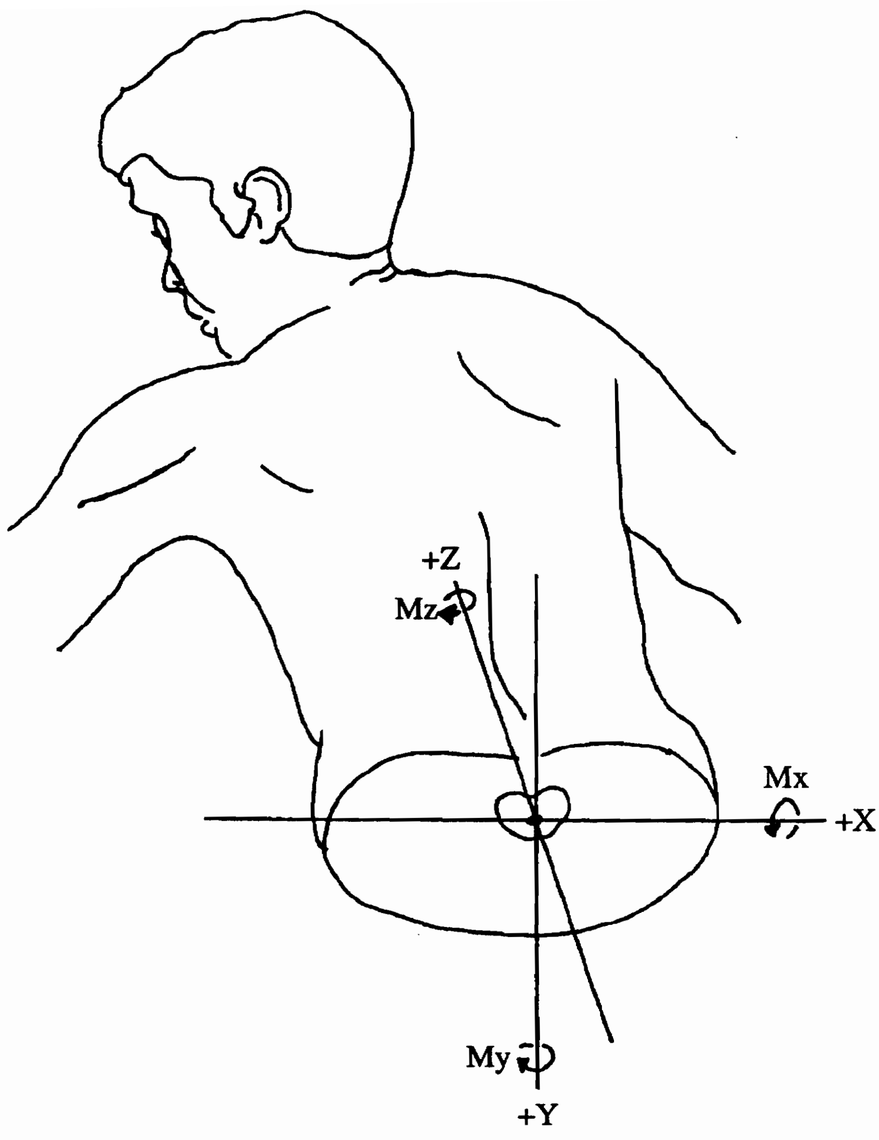

Figure 2. Three-dimensional coordinate system used in the model. Adapted from Hughes (1991).

The model first assumes that the Z - axis of the body is represented by a vector from the midpoint of the knees to the L3/L4. This origin is then translated to the L3/L4 location. This vector approximates the orientation of the torsional axis of the L3/L4 joint. This vector was still expressed in terms of the original (room) coordinate system.

The next stage in the transformation process was to define a plane orthogonal to the Z - axis of the body. The *Gram-Schmidt orthogonalization process* was used to define this plane. A good discussion of this procedure can be found in Bradley (1975). Two vectors, γ_2 and γ_3, which are both orthogonal to the Z - axis, are computed as:

$$\gamma_2 = \alpha_2 - \left[\frac{(\alpha_2|\gamma_1)}{(\gamma_1|\gamma_1)}\right]\gamma_1 \qquad (1.10),$$

$$\gamma_3 = \alpha_3 - \left[\frac{(\alpha_3|\gamma_1)}{(\gamma_1|\gamma_1)}\right]\gamma_1 - \left[\frac{(\alpha_3|\gamma_2)}{(\gamma_2|\gamma_2)}\right]\gamma_2 \qquad (1.11),$$

where γ_1 represents the Z - axis expressed as a vector. The variables α_2 and α_3 are defined as any two vectors that are not scalar multiples of γ_1, γ_2, or γ_3. As part of the experimental measurements, a vector was established from the L3/L4 joint to a point located approximately 11 cm lateral of L3/L4. This vector was measured to establish to what degree the torso cutting plane had rotated about L3/L4. This point was not necessarily in the plane orthogonal to the Z - axis as defined above (see in Figure 3). To account for this, the X - axis was defined by projecting the vector from L3/L4 to the lateral torso point onto the orthogonal plane defined by γ_2 and γ_3. The X - axis can then be expressed as

$$\beta = \left[\frac{(\alpha|\gamma_2)}{(\gamma_2|\gamma_2)}\right]\gamma_2 + \left[\frac{(\alpha|\gamma_3)}{(\gamma_3|\gamma_3)}\right]\gamma_3 \qquad (1.12),$$

where α represents the L3/L4 lateral vector and γ_2 and γ_3 are as before.

The Y - axis was then derived by calculating a right-hand vector orthogonal to both γ_1 (the Z - axis) and β (the X - axis). The Y- axis can be expressed as

$$\delta = \gamma_1 \times \beta \qquad (1.13),$$

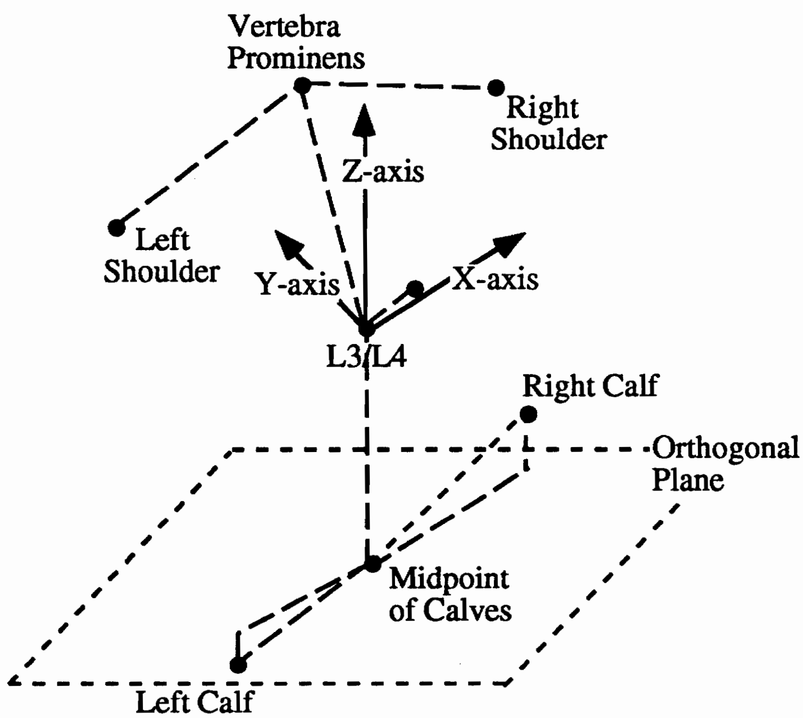

Figure 3. The axes of the body coordinate system as defined by the vectors β, δ, and γ_1 (X, Y, and Z - axes, respectively).

or the cross product of γ_1 and β. Figure 2.3 illustrates the different axes as defined above.

With the axes that compose the body reference frame defined, it was necessary to create a transformation matrix that could be used to express vectors measured in the room reference frame in terms of the body reference frame. A good discussion of this process can be found in Craig (1989). The axis vectors defining the body reference frame were converted into unit vectors and then broken into their individual components. If **x**, **y**, and **z** are defined as the unit vectors of β, δ, and γ_1 respectively, then the individual components of these unit vectors can be expressed as:

$$\begin{aligned}\mathbf{x} &= (a_1, a_2, a_3),\\ \mathbf{y} &= (b_1, b_2, b_3),\\ \mathbf{z} &= (c_1, c_2, c_3),\end{aligned} \qquad (1.14),$$

and a 3 x 3 transformation matrix, ${}^{\text{room}}_{\text{body}}\mathbf{Trans}$, was constructed from these individual unit vector components as follows:

$$ {}^{\text{room}}_{\text{body}}\mathbf{Trans} = \begin{vmatrix} a_1 & a_2 & a_3 \\ b_1 & b_2 & b_3 \\ c_1 & c_2 & c_3 \end{vmatrix} \qquad (1.15).$$

Any vector expressed in terms of the room reference frame can then be transformed to the body reference frame by multiplying the transformation matrix as follows:

$$ {}^{\text{body}}\mathbf{V} = {}^{\text{room}}_{\text{body}}\mathbf{Trans}\ {}^{\text{room}}\mathbf{V} \qquad (1.16),$$

where ${}^{\text{body}}\mathbf{V}$ is the vector expressed in the body reference frame and ${}^{\text{room}}\mathbf{V}$ is the vector expressed in the room reference frame.

Transformation matrices calculated as described above were used to convert the resultant moment and force vectors (Equations 2.7 and 2.9) to the body reference frame. These moments were then used as inputs to the optimization algorithms as described below. The force and posture measurement apparatus are described in greater detail in the next chapter.

Internal Muscle Forces

The double linear programming (DLP) method of Bean et al. (1988) was used to estimate muscle forces and compressive force at the L3/L4 joint. This approach estimates the muscle forces required to maintain static equilibrium, and which minimize the maximum muscle intensity and then the intervertebral disc compression. The analysis was performed at an imaginary cutting plane at the L3/L4 level, which is equivalent to the orthogonal plane described in the previous section. Figure 4 shows the five left-right muscle pairs which were included in the model: erector spinae (LES, RES), latissimus dorsi (LLD, RLD), rectus abdominus (LRA, RRA), external obliques (LEO, REO), and internal obliques (LIO, RIO). Muscle moment arms, cross-sectional areas, and lines-of-actions for the ten muscles were required for the formulation. The muscle moment arms and cross-sectional areas of Tracy, Gibson, Szypryt, Rutherford, and Corlett (1989) and the muscle lines of action of Dumas, Poulin, Roy, Gagnon, and Jovanovic (1988) were used in the model and are shown in Table 2.

The requirement that the moments due to internal forces must balance the moments due to externally applied forces, can be expressed as:

$$\sum_{i=1}^{n} \|\mathbf{f}_i\| \left(\mathbf{r}_i \times \mathbf{t}_i\right) + \mathbf{M}_{ext} = 0 \qquad (1.17),$$

where $\mathbf{M}_{ext}$ is the three-dimensional moment due to externally applied forces, and f_i is the three-dimensional force vector for muscle i. The variable n represents the number of muscles included in the analysis (n=10 in this case). In applying the Bean et al. (1988) DLP algorithm to the L3/L4 joint, the first step was to minimize the maximum muscle contraction intensity:

$$\begin{aligned} &\text{minimize } \mathrm{I} \qquad (1.18),\\ &\text{subject to: } \sum_{i=1}^{n} \|\mathbf{f}_i\| \left(\mathbf{r}_i \times \mathbf{t}_i\right) = -\mathbf{M}_{ext}\\ &\frac{\|\mathbf{f}_i\|}{A_i} \le \mathrm{I},\\ &\|\mathbf{f}_i\| \ge 0, \end{aligned}$$

where **f** represents the force in muscle i and **r** and **t** are three-component vectors representing the x, y, and z directions of moment arm and unit force, respectively. The variable $\mathbf{M}_{ext}$ represents the three-component moment caused by the externally applied forces in the body reference frame (see above). Because multiple optima may exist for the first linear program (Eq. 2.18), the model employs a second step to derive the single optimum which also minimizes the compressive forces. This linear program is as follows:

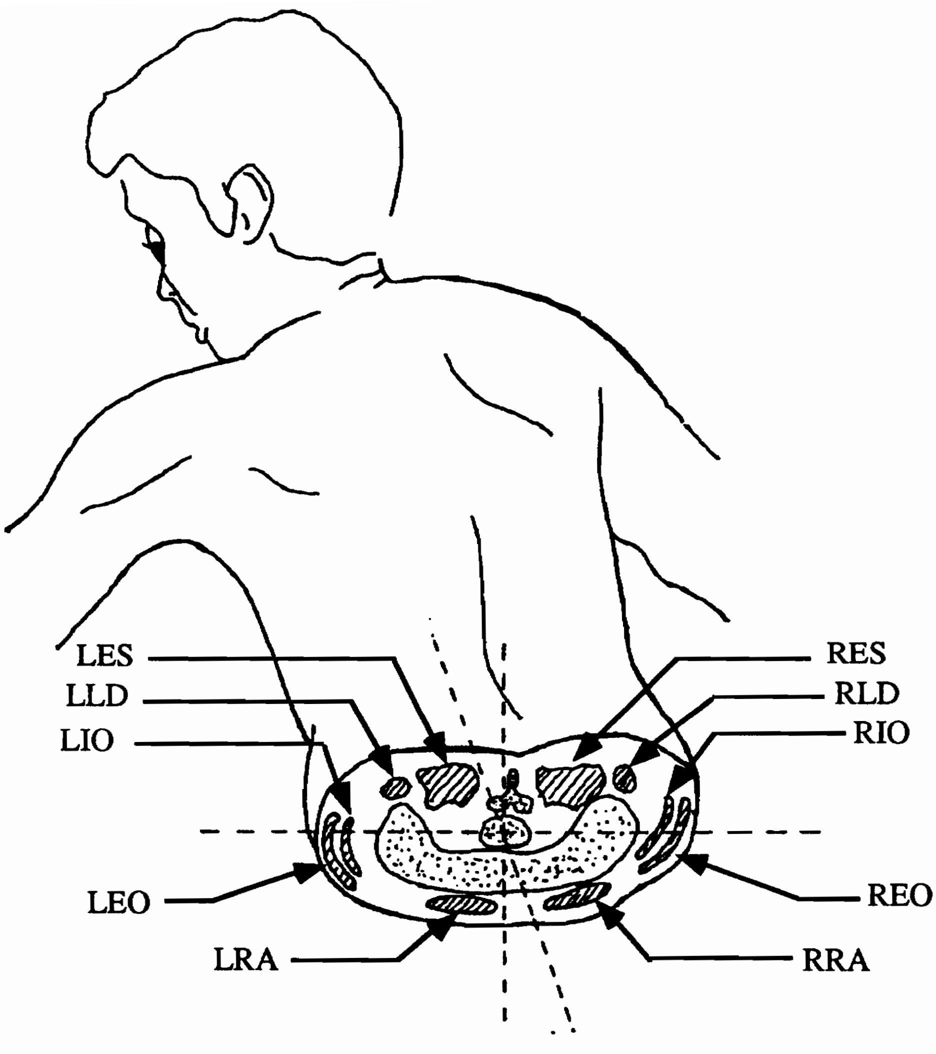

Figure 4. The five lumbar muscle pairs used in the model: Erector Spinae (LES, RES), Latissimus Dorsi (LLD, RLD), Internal Obliques (LIO, RIO), External Obliques (LEO, REO), and Rectus Abdominus (LRA, RRA). Adapted from Hughes (1991).

Table 2. Data Used in Optimization Model: Moment Arms R_x, R_y, R_z (cm), Unit Muscle Force Components T_x, T_y, T_z, and Anatomical Cross-Sectional Areas A_i (cm^2), Adapted from Hughes (1991)

Muscle	r_x^i	r_y^i	r_z^i	t_x^i	t_y^i	t_z^i	A_i
L. Erector Spinae	-3.82	-5.76	0.0	0.281	-0.052	-0.9580	24.
R. Erector Spinae	3.82	-5.76	0.0	-0.281	-0.052	-0.9580	24.
L. Rectus Abdominus	-3.38	7.95	0.0	-0.028	0.016	-0.9995	6.6
R. Rectus Abdominus	3.38	7.95	0.0	0.028	0.016	-0.9995	6.6
L. Internal Oblique	-11.39	0.96	0.0	-0.134	-0.574	-0.8080	14.
R. Internal Oblique	11.39	0.96	0.0	0.134	-0.574	-0.8080	14.
L. External Oblique	-12.62	1.09	0.0	0.376	0.322	-0.8700	15.
R. External Oblique	12.62	1.09	0.0	-0.376	0.322	-0.8700	15.
L. Latissimus Dorsi	-7.19	-5.42	0.0	0.340	-0.284	-0.8970	3.9
R. Latissimus Dorsi	7.19	-5.42	0.0	-0.340	-0.284	-0.8970	3.9

$$\text{minimize} \sum_{i=1}^{n} \|\mathbf{f}_i\| t^i{}_z \qquad (1.19),$$

$$\text{subject to:} \quad \sum_{i=1}^{n} \|\mathbf{f}_i\| (\mathbf{r}_i \times \mathbf{t}_i) = -\mathbf{M}_{ext}$$

$$\frac{\|\mathbf{f}_i\|}{A_i} \leq I^*,$$

$$\|\mathbf{f}_i\| \geq 0,$$

where I^* represents the maximum intensity found as a result of the first optimization step.

Finally, compressive force at the L3/L4 disc was computed for the optimal solution by adding all lumbar muscle forces in the z direction (perpendicular to the cutting plane) and the resultant force due to external forces as follows:

$$F_{Compression} = \sum_{i=1}^{n} \|\mathbf{f}_i\| t^i{}_z + F_{L3/L4}z \qquad (1.20).$$

In this formula, $F_{L3/L4}z$ represents the z component of the external force acting on L3/L4 in equation 2.8.

All components of the biomechanical model were implemented as Mathematica programs on an Apple Macintosh computer.

VITA

Brian Randall Sherman was born on July 17, 1969 in Durham, North Carolina. He received his B.S. in Industrial and Systems Engineering at The Ohio State University, Columbus, Ohio in June of 1992. He went on to pursue his graduate work in the Industrial Ergonomics Laboratory in the Human Factors Engineering Center at Virginia Polytechnic Institute and State University, Blacksburg, Virginia. He is a active member of the Human Factors and Ergonomics Society, American Society of Safety Engineers, and the Institute of Industrial Engineers. Brian will be working for Sumitomo Electric Wiring Systems, Incorporated as an ergonomics engineer at their components manufacturing plant in Scottsville, Kentucky.

B. Randall Sherman

Brian Randall Sherman