KINETIC AND KINEMATIC ADAPTATIONS TO USE OF A PERSONAL LIFT ASSIST DEVICE

by

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Abstract

The purpose of this work was to quantify the effect of the personal lift assist device (PLAD) on kinetic and kinematic variables commonly indentified as risk factors for low back pain (LBP). As such, three investigations were undertaken to document adaptations that occur as a consequence of wearing the PLAD.

The first study involved an investigation of the effects of the PLAD on intervertebral compression and shear, using an EMG-assisted biomechanical model across a range of trunk flexion. Muscle activation (EMG), trunk posture, and PLAD support data were input into a biomechanical model that estimated L4/L5 joint loads. Use of the PLAD significantly reduced joint compression across the range of trunk flexion. Significant changes in shear were also found, although this was varied across conditions.

The second study was conducted to quantify differences in lifting posture, lifting kinematics, and co-ordination attributable to use of the PLAD. Over two testing sessions, subjects completed a repetitive lifting task with and without use of the PLAD. Kinematic data describing lumbar spine, hip and knee motion were used to quantify lifting posture, lifting velocities, and co-ordination. The results of the study suggest that the PLAD causes users to lift with significantly less lumbar flexion and greater hip flexion. Significant changes in co-ordination were also observed, reflective of motor adaptation to the assistance provided by the PLAD.

The final study was conducted to investigate the effect of the PLAD on active trunk stiffness. Subjects were required to assume a series of static, symmetrical flexed postures. Muscle activation (EMG), trunk posture, and PLAD stiffness data were input into a stability model that estimated active trunk stiffness. Up to 15 degrees of flexion, the PLAD increased the overall stiffness of the trunk. However, use of the PLAD significantly reduced the active

stiffness of the trunk as flexion increased. This effect was consistent across PLAD conditions. Further research is needed to confirm these findings and evaluate a potential redesign. In general, the results of these studies illustrate the potential for the PLAD to be used as an ergonomic intervention for industrial tasks requiring lifting and/or flexed static postures.

Co-Authorship

The manuscripts presented in Chapters 2-4 are in the process of being submitted for publication in peer-reviewed journals. The author listings for these works are as follows:

1. Agnew, M.J., Brown, S.H.M., Wilkes, J., and Stevenson, J.M. Reducing spinal loads through use of a Personal Lift Assist Device (PLAD): Effects of trunk flexion and level of PLAD support.

2. Agnew, M.J., Godwin, A.A., and Stevenson, J.M. The impact of a Personal Lift Assist Device (PLAD) on lifting kinematics and co-ordination.

3. Agnew, M.J., Brown, S.H.M., and Stevenson, J.M. The effect of a Personal Lift Assist Device (PLAD) on active trunk stiffness: Implications for spine stability.

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Statement of Originality

I hereby certify that all of the work described within this thesis is the original work of the author. Any published (or unpublished) ideas and/or techniques from the work of others are fully acknowledged in accordance with the standard referencing practices.

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Chapter 1
Introduction

Despite being formally documented by Ramazzini (1713) nearly 300 years ago, occupationally related musculoskeletal injuries, better known as work-related musculoskeletal disorders (WMSDs), continue to manifest in working populations spanning a broad spectrum of occupational environments (Fathallah et al., 2004). Of these disorders, low back pain (LBP) is reported to be the most common musculoskeletal problem reported by people seeking medical care (Oddsson et al., 1997). While it has been reported that the relative incidence of occupational low back pain is on the decline (Murphy & Volinn, 1999), the associated costs and impact of LBP remain significant, from both a societal and economic context. In developed, industrialized countries, the estimated lifetime prevalence for LBP is reported to range between 49-84% (van Tulder, Koes, & Bombardier, 2002); within United States, the yearly financial impact associated with this prevalence of LBP is estimated to exceed $26 billion (Luo et al., 2004).

One of the most problematic aspects regarding the development of LBP stems from the fact that the causes of LBP are multi-factorial and have been demostrated to synergistically interact (Marras et al, 1995). While an exhautive list of significant risk factors has been published in the literature, they are typically characterized as being personal, psychosocial, or physical in scope (Davis & Jorgensen, 2005). In terms of physical causes for LBP, epidemiological research suggests that occupational risk factors, namely physical workplace exposures, account for 30-40% of reported LBP episodes and injuries (Norman et al., 1998; Punnet et al., 2005). Example risk factors include: awkward postures and motions, prolonged static flexion, heavy physical work and lifting, forceful movements. and peak and cumulative biomechanical joint loads (Chaffin and Park, 1973; Bigos et al., 1986; Kumar, 1990; Marras et al., 1993 & Norman et al., 1998).

Using this knowledge, ergonomists continually strive to reduce the presence and/or severity of physical risk factors found within a given workplace. Whenever possible, tasks identified as physically demanding are redesigned to eliminate the associated injury risks. In the event this is not possible, task demands are often minimized through the use of material handling devices designed to assist the operator. However, many occupational settings exist where workspace redesigns or implementation of material handling devices cannot be facilitated or cost justified. In the event that workstation or process redesign is not possible, alternate ergonomic control strategies are sought in order to minimize the biomechanical demands and injury risks associated with a given task.

In recent years, a number of ergonomic assistive devices have been made commercially available, and research has documented their effectiveness at reducing low back loads during occupational tasks. For example, in an analysis of peak and cumulative spine loads experienced by sheep shearers, Gregory et al. (2006) documented the effectiveness of a supportive harness in reducing peak and maximum spine loads incurred during sheep shearing tasks. More recently, Paskeiwicz & Fathallah (2007) quantified the effectiveness of the GRIPSystem™, an assistive device designed to reduce spinal loads and LBP risk in furniture handling tasks. Both of these studies support the notion that passive, assistive devices can be used to successfully reduce the physical demands placed on the spine during material handling tasks.

Over the last five years, our lab has been conducting similar research regarding the effectiveness of an assistive device. We have channeled our efforts toward the development of an ergonomic lifting aid known as the Personal Lift Assist Device (PLAD). It is designed to reduce the physical demands and biomechanical loads placed on the tissues of the low back during tasks requiring manual material handling and tasks requiring prolonged trunk flexion.

The PLAD system is modeled on the concept of the passive elastic properties of human muscles. As such, the PLAD features an external elastic element that runs along a relatively similar line of action to that of the erector spinae musculature (Abdoli-Eramaki, 2005). Acting through a 0.12 m lever that projects dorsally from a belt worn around the waist, the PLAD element inherently possesses a mechanical advantage over the extensor musculature that span the lumbar vertebrae. During lifting tasks and static postures, flexion of the trunk stretches the PLAD elastic elements, creating an assistive extensor moment about the low back and hips. This assistive moment reduces the physical demand placed on the trunk extensor musculature, ultimately leading to a decrease in muscle activation and force. The past severeal years have seen several design changes in the PLAD; these are illustrated in Figure 1-1.

Figure 1-1 Comparison of PLAD design iterations used within previous experiments.
Left: PLAD version 1 - in Abdoli et al., 2006, 2007, 2008. **Right**: PLAD version 3 - in Lotz et al., 2008, Godwin et al., (in press), and the research presented in this thesis.

In previous iterations, the PLAD featured four elastic elements that ran from the shoulder attachment points to the PLAD lever arm and two elastic elements that connected the lever arm to the knee attachment points. In successive design changes, the number of elastic elements was reduced, and the PLAD elastic element length was shortened and attached to cables. In addition, pulleys were fixed to the PLAD lever arm in order to guide the cables running from the knees to the PLAD elastic elements. While the number and size of elements changed from PLAD version 1 to PLAD version 3, the effective PLAD moment arm about the flexion/extension axis remained the same. However, the element stiffness was changed slightly in order to ensure that PLAD version 3 provided similar extensor moments as PLAD version 1 for a given amount of trunk flexion.

Regardless of design iteration, the effectiveness of the PLAD at providing assistive support to the extensor musculature has been consistently observed. In simulated lifting tasks, the moment-generating capacity of the PLAD has been shown significantly reduce the average muscula activity (EMG) of the thoracic and erector spinae musculature, with no concurrent increase in trunk flexor coactivity. This response has been observed both in male and female subjects, and in tasks involving both symmetrical and asymmetcial postures (Abdoli et al., 2006; Stevenson et al., 2007; Abdoli & Stevenson, 2008). Furthermore, in repetitive lifting studies testing male and female subjects, use of the PLAD has been found to prevent the onset of localized muscle fatigue in the thoracic and lumbar erector spinae; this has been evidenced through amplitide and spectral analyses of thoracic and lumbar EMG signals (Lotz et al., 2007; Godwin et al., in press).

Collectively, the results of these studies illustrate the effect of the PLAD at reducing the physical demands placed on the erector spinae during lifting tasks. However, many questions remain unanswered in terms of the effect that the PLAD has on known kinetic- and kinematic-based risk factors for LBP, including: spinal loads (compression and shear), lifting postures, and trunk stiffness. While results from previous research indirectly imply that using the PLAD should have an effect on these measures, these hypthotheses have not been formally tested to date.

The aim of the current work was to quantify the effect of the PLAD on select kinematic and kinetic lifting variables that have been indentified as potential risk factors for low back pain (LBP). Specifically, the current work attempted to quantify:

1. The effect of the PLAD device on lifting kinematics, trunk postures, and co-ordination between segments.
2. Associated changes in L4/L5 intervetebral compression and shear forces attributable to use of the PLAD.
3. The effect of the PLAD on active trunk stiffness during static trunk flexion.

It was anticipated that the results of the study would provide a comprehensive assessment of the kinetic and kinematic adaptations that occur due to use of the PLAD device. As such, the results of the study lend futher evidence to the notion that the PLAD is a safe and effective ergonomic intervention, suitable for implementation within the workplace. Additionally, the results of the study will also be used as a basis for future research designed to assess whether design changes are necessary in order to improve the effectiveness of the PLAD system.

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Chapter 2
Reducing Spinal Loads through Use of a Personal Lift Assist Device (PLAD): Effects of Trunk Flexion and Level of PLAD Support

2.1 Abstract

The associated risks that excessive peak and cumulative spinal loads have on the development of low back pain are well known. Despite the establishment of exposure guidelines, occupationally related low back injuries continue to manifest and warrant alternative control strategies. Over the past few years, we have been developing an on-body lift assist device called the Personal Lift Assist Device (PLAD). This system is worn by workers and features elastic elements that stretch during trunk flexion to create an assistive moment about the trunk. The purpose of the study was to quantify the effects of the PLAD on intervertebral compression and shear loads, using an EMG-assisted biomechanical model. Nine male subjects participated in the study. Subjects were required to assume a series of static, symmetrical flexed postures, varied by trunk flexion magnitude. Within each posture, a No-PLAD condition was tested, as were five different levels of PLAD support. Normalized electromyographic (EMG) data were quantified across seven muscles on the right side of the body with bilateral symmetry assumed. These data, in addition to trunk posture, and PLAD support data were input into an EMG-assisted biomechanical model that estimated L4/L5 intervertebral joint loads based on the input data and anatomical and geometric constraints. Use of the PLAD system was shown to significantly reduce joint compression across the range of trunk flexion by magnitudes ranging from 191 to 646 N. Significant changes in joint shear force were also found, although the effect was varied across conditions. The results of the study further illustrate the potential benefits of the PLAD as a means of reducing peak spine and cumulative spine loads. Future work should attempt to assess the effectiveness of the device with asymmetric tasks and more complex biomechanical loads.

2.2 Introduction

Of all work-related musculoskeletal disorders (WMSDs), low back injuries are the most common type of injury (Marras, 2000). Within the United States alone, the annual financial impact associated with low back pain (LBP) is estimated to exceed $26 billion (Luo et al., 2004). Epidemiological LBP research suggests that occupational risk factors, namely physical workplace exposures, account for 30-40% of reported injuries (Norman et al., 1998; Punnet et al., 2005). Example risk factors include: heavy physical work and lifting, forceful movements. and peak and cumualtive biomechanical joint loads (Chaffin and Park, 1973; Kumar, 1990; Marras et al., 1993 & Norman et al., 1998).

Applying this knowledge, ergonomists devise and implement solutions that attempt to minimize the risk level associated with a given workstation or process that involves any of the previously mentioned types of physical demands. The definition of "risk" within a given occupational environment is usually made through quantification of a task's physical demands and its subsequent comparison to a recommended guideline, the most recognizable being the NIOSH *Guide to Manual Materials Handling* and its *Lifting Equation*: an evaluation metric believed to be based on epidemiological, physiological, psychophysical, and biomechanical evidence associated with LBP risk (Waters et al. 1993). When risks of injury are deemed present, an attempt is then made to improve the workstation, or process, by eliminating the factors that are posing the most risk. However, many occupational settings exist where redesigns cannot be facilitated or cost justified, and alternative solutions are required in order to reduce the physical exposures placed on workers.

Over the past few years, researchers in our lab have been developing an on-body lift assist device called the Personal Lift Assist Device (PLAD). This system is worn by workers and features elastic elements that stretch during trunk flexion to create an assistive moment about the trunk. Based on the mathematical proof provided in Abdoli et al. (2007), the PLAD elastic elements have a mechanical advantage over the erector spinae musculature. Accordingly, empirical research has documented the assistive support provided by the device, as comparative erector spinal activation levels have been shown to decrease during lifting tasks using the PLAD (Abdoli et al., 2006). In theory, this should decrease the net joint compression and shear forces experienced at the L4/L5, provided that co-contraction of the flexor musculature remains relatively unchanged and does not increase in opposition to the extensor moment (elastic resistance) caused by the PLAD. The purpose of the current study was to further investigate the notion that use of the PLAD could reduce spinal loads incurred during occupational tasks requiring lifting or static flexion of the trunk. This was achieved through use of an EMG-assisted biomechanical model sensitive to activation changes in both the flexor and extensor musculature that support the trunk.

2.3 Methods

2.3.1 Participants

Nine healthy male subjects volunteered to participate in the study. Subjects were from a university population and had a mean (SD) age of 21.7 (0.9) years, height of 1.81 (0.05) m and a mass of 81.3 (8.1) kg. On the day of testing, all subjects reported that they were free of known skeletal muscle impairment and without any previous back, hip, or knee problems. Subjects were required to sign informed consent (as approved by the Queen's University Research Ethics Board) and complete a PAR-Q Health questionnaire (Appendix B).

2.3.2 Instrumentation

2.3.2.1 EMG Data

Bipolar Ag/AgCl EMG electrodes (Medi-Trace™) were placed on the skin, with an inter-electrode distance of 2.5 cm, at seven different locations on the right side of the body in order to record signals from the following muscles: rectus abdominus (RA), external oblique (EO), internal oblique (IO), latissimus dorsi (LD), thoracic erector spinae (TES), lumbar erector spinae (LES), and multifidus (MULT). Prior to the attachment of the electrodes, the skin over the muscle was cleaned and abraded with alcohol. Electrodes were placed according to the procedures described in Cholewicki & McGill (1996). Recordings of the TES and LES were taken at the T9 and L3 levels, with the electrodes laterally placed 5 cm and 3 cm from the respective spinous processes. Electrodes monitoring MULT activity were placed 2 cm lateral to the L4/L5 spinous process. The LD electrodes were placed lateral to T9 over the muscle belly. RA electrodes were placed 3 cm lateral to the umbilicus, while EO electrodes were placed 15 cm lateral to this landmark at an oblique angle along the muscle's line of action. Finally, IO electrodes were placed approximately midway between the anterior superior iliac spine and the symphysis pubis, above the inguinal ligament. A reference electrode was placed over the C7 spinous process. Bilateral symmetry was assumed, and right side signals were used to represent both the ipsi- and contralateral musculature. The assumption of equivalent right/left muscle activations was deemed valid, as this response has been identified in previous research involving sagittal lifting (Potvin et al., 1996; Granata et al., 1997), including investigations involving use of the PLAD (Abdoli et al., 2006).

Muscle signals were subsequently scaled with respect to EMG amplitudes obtained from a series of maximum voluntary contractions (MVC). Collection of these data required subjects to maximally contract while adopting a series of postures designed to elicit true maximum efforts for each of the seven muscles. Back extensor MVCs (TES, LES, MULT) were collected while the subject lay prone on a table, secured at the pelvis and legs. Subjects were then required to perform a series of isometric extension exertions against resistance provided by the experimenter. Flexor MVCs required the subjects to lie supine with their knees bent and ankles flat on the testing table. For the RA, subjects were required to perform an isometric abdominal crunch against the resistance of the experimenter. For the remaining two flexor muscles (EO, IO), subjects were again required to perform an abdominal crunch; however subjects were asked to also couple the crunch with a slight twisting motion (left and right) in accordance with the oblique action lines of both the EO and IO musculature. For the final test, the subject sat on the bench and performed a resisted "pull-down" exercise targeted to activate the LD; they sat with their upper arm abducted and elbow flexed to 90 degrees and pushed down while the experimenter resisted the downward motion. Each contraction was held for 3-5 seconds and adequate rest was provided between tests. Generally, three repetitions were required for each muscle group; however, in some cases, additional contractions were necessary in order achieve maximum efforts.

Throughout the maximum exertion trials and test conditions, EMG signals were amplified (gain = 1K-5K; input impedance = 10GΩ; CMRR = 115 dB at 60Hz, Bortec Octopus AMT-8, Calgary, Canada) and digitally captured using a 12-bit A/D card (National Instruments, Austin, TX, USA) at 1024 Hz. The raw EMG data were then digitally band-pass filtered (30-450 Hz) and stored for post processing. Recent research has recommended a lower cut-off of 30 Hz in order to remove heart rate artifact known to contaminate trunk EMG signals (Drake & Callaghan, 2006).

2.3.2.2 PLAD Device and PLAD Elements

Five different elements of varying stiffness were created for use with the PLAD during this study. The elements were made out of the similar elastic material to that used in previous design iterations (JumpStretch™ as cited in Frost et al. (2006)). The length of the elements was determined based on consultation with an expert who suggested that elastic materials, such as those used within the PLAD, respond most reliably through a strain of 50-100% (Riazi, personal communication). Based on these suggestions and the expected elongation of the elements as reported in Abdoli et al. (2007), each element was made to be 20 cm in length. Additionally, the five elements were also designed to have an elastic stiffness that would generate forces ranging between 50-150% of what has been observed in previous investigations of the device (Abdoli et al., 2006).

2.3.2.3 PLAD Load Cell

The force generated by the PLAD elements was monitored using a load cell placed in series with one of the PLAD elements, between the shoulder and hip attachment points on the right side of the device. The load cell was custom-made and consisted of a strain gauge attached to an aluminum plate. Prior to testing, the load cell was calibrated using a set of known masses and use of a linear regression function, as the response of the load cell was found to be consistently linear (R^2 = 0.90 to 0.98 over repeated tests). During the experiment, voltages from the load cell were appropriately gained using a DC Amplifier (RDP Group, Modular 600), digitally sampled (along with the EMG data) at 1024 Hz using a 12-bit A/D card (National Instruments, Austin, TX) and stored for post processing. During post processing, force signals were doubled in amplitude to estimate the combined forces in both the right- and left-side elastic elements, and thus the total force-generating capacity of the device.

2.3.2.4 Motion Analysis Hardware

Trunk flexion was monitored using a Fastrak™ electromagnetic motion analysis system (Polhemus, Colchester, VT). Prior to the trial conditions, Fastrak™ motion sensors were placed on the sternum and over the left iliac crest and firmly attached using TUF-Skin® spray and adhesive tape. Tubular bandages (Mefix™) were tied around the chest to assist in securing the Fastrak™ sensor. Fastrak™ data were normalized to an anatomical reference posture using the "boresight" function within the Fastrak™ hardware, so that an estimate of trunk flexion angle could be determined. The relative angular displacement between the sensors was used to represent flexion across the lumbar and thoracic regions of the spine. These data were collected in real time (30Hz) and presented to the subjects and experimenters via a graphical display created using custom software (LabVIEW 7.1, National Instruments, Austin, TX).

2.3.3 Experimental Protocol

Testing required subjects to perform 30 trials varied by flexion angle and PLAD stiffness. When ready, subjects were required to bend forward through their trunk range of motion to one of five angles (15, 30, 45, 60, or 75 degrees). Biofeedback (trunk flexion) was presented to subjects on a computer monitor, thus enabling them to position themselves at the appropriate flexion angle. In addition, subjects were instructed to let their arms hang freely from their shoulders while holding on to a small (0.20m*0.15m*0.1m), empty cardboard container. This postural constraint was included for consistency in the location of the trunk and the upper limbs across all tasks and subjects. For each posture, subjects were required to wear the PLAD device, configured to one of six predetermined levels of assistive support; the six levels of PLAD stiffness included a No-PLAD condition as well as five PLAD conditions (K1-K5) of increasing support. Subjects wore the PLAD device during the No-PLAD condition; however, the shoulder and hip components were disconnected, preventing the device from having any mechanical effect.

Trial order was randomized in blocks of element stiffness. Within each block, flexion angle was randomized as well. Data collection began once the subject reached the required flexion angle and remained motionless (as determined from the biofeedback); the subject was then instructed to maintain this posture for the five seconds of data collection.

A scaled digital photograph was taken immediately following each experimental condition while the subject remained in a flexed posture. To reduce the likelihood of fatigue onset, short rest periods between each trial were given, along with longer breaks that occurred during the changing of the PLAD elements. Between each trial, the experimenter visually inspected the PLAD to ensure that it was positioned properly at the shoulders, hips, and knees; adjustments were made when necessary. Postural, biomechanical, and EMG data collected during these trials were input into an EMG-assisted spine model to calculate L4/L5 compression and anterior/posterior (A-P) shear. A graphical depiction of the experimental protocol and instrumentation is presented in Figure 2-1.

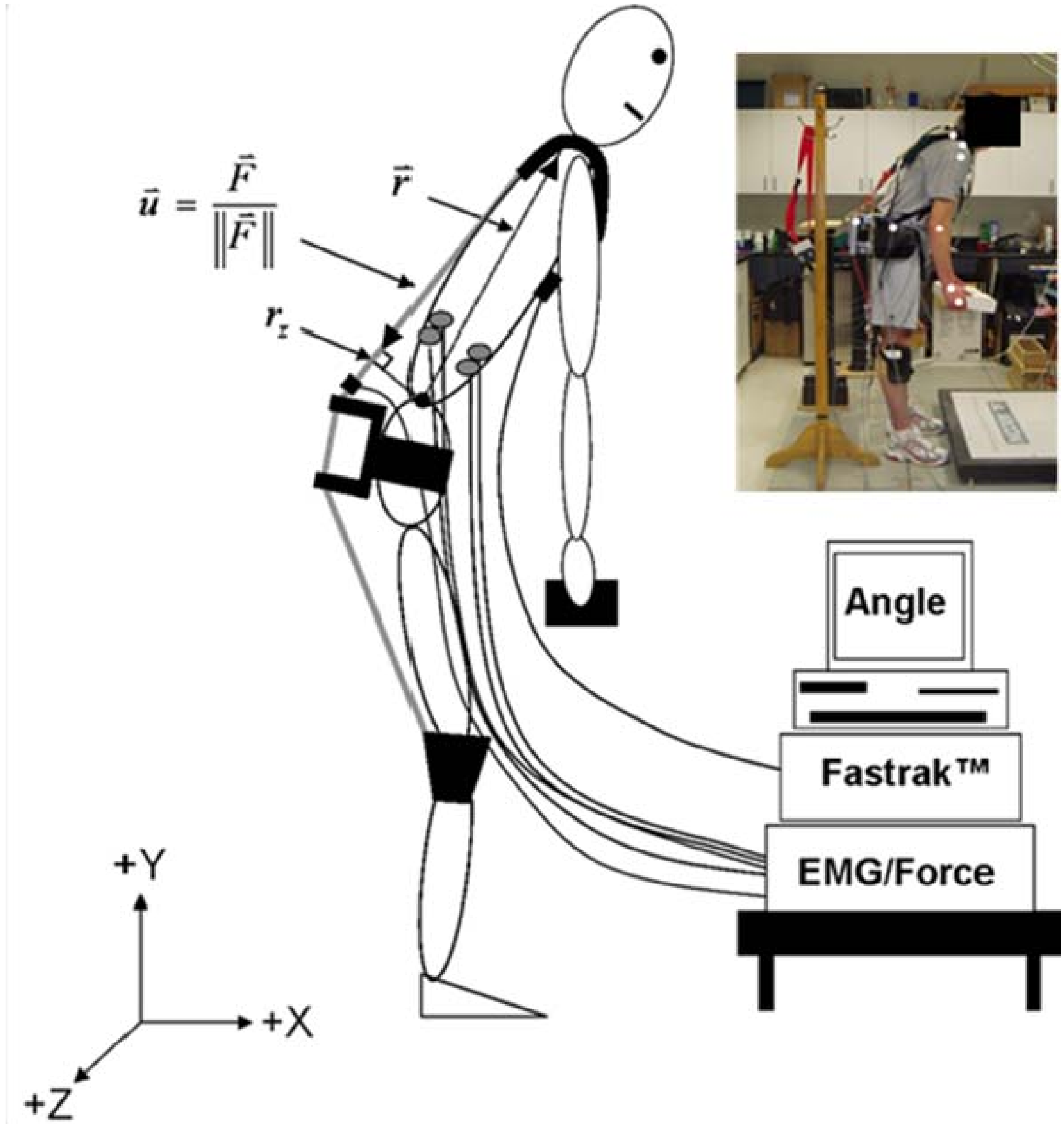


Figure 2-1 Graphical representation of the experimental protocol (photo inset) used within the study. The variables F, u, r, and r_z represent PLAD force, PLAD force unit vector, PLAD element insertion relative to L4/L5, and the PLAD moment arm respectively about the flexion-extension axis.

2.3.4 Data Processing

2.3.4.1 EMG-Assisted Biomechanical Model

Intervertebral joint loads (L4/L5 level) were predicted using an EMG-assisted biomechanical model (Brown & Potvin, 2005). The anatomically detailed model was representative of a 50th percentile male; and consisted of a pelvis, five lumbar vertebrae, and a ribcage. Global, three-dimensional joint co-ordinates, muscle attachment points, and physiological cross-sectional areas used within the model were taken from the data of Cholewicki & McGill (1996). As the focus of the study was to assess L4/L5 joint loads, the number of muscles used in the analysis was reduced to only include fascicles that crossed the L4/L5 joint. In total, 58 (29 right, 29 left) muscle fascicles were subsequently included in the model, representing the following musculature (# of fascicles); Rectus Abdominus (RA–2), External Oblique (EO-4), Internal Oblique (IO-4), Thoracic Erector Spinae (TES-6), Lumbar Erector Spinae (LES-8), Multifidus (MULT-14), Latissimus Dorsi (LD-4), Quadratus Lumborum (QL-8), and Psoas (PS-8). Each muscle fascicle included an active contractile and passive elastic component. Model co-ordinates are presented graphically in Figure 2-2.

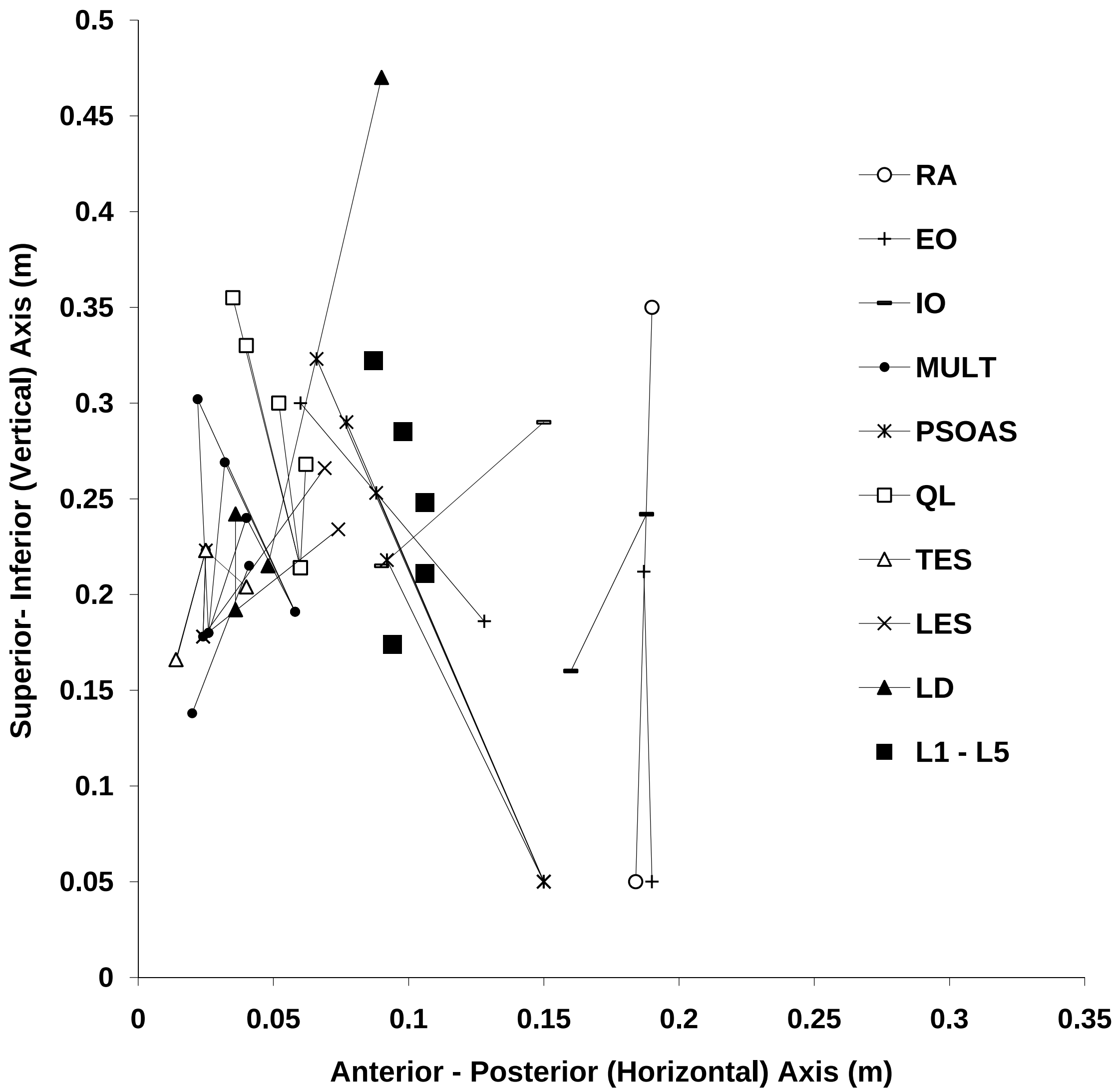


Figure 2-2 Sagittal view of muscle origins, insertions, and vertebral co-ordinates used within the EMG-based biomechanical model of the lumbar spine. Data are expressed relative to a global co-ordinate system located outside of the body, as defined by Cholewicki and McGill (1996).

2.3.4.2 Model Inputs

The model required three inputs: Euler angles describing the three-dimensional (3D) orientation of the trunk, net moments acting about the joint(s) of interest (in this case L4/L5), and EMG signals to assist in the prediction of force within each fascicle. As this study examined symmetrical postures varying in trunk flexion, the orientation of the trunk and the external demands were described with a single rotation angle (trunk flexion) and an estimate of the external flexion moment (M_z) acting at the L4/L5 joint. Using the trunk angle data, rotations were assigned to each intervertebral joint. These rotations were assumed to be proportional across vertebrae and were partitioned using McGill & Norman (1986):

$$R_{(i)} = \theta_T * L * \psi_{(i)} \qquad [1]$$

Where $R_{(i)}$ = rotation of the ith vertebra (rads), (i) = lumbar level (L1-L5), θ_T = trunk flexion (rads), L = proportion of trunk flexion occurring in the lumbar spine (set to 0.513 based on data from White & Panjabi (1990), and $\psi_{(i)}$ = % flexion assigned to each vertebra (L1-L2: 13.2%, L2: 21%, L4: 29%, L5: 23.6%) .

Using these rotations, subsequent changes in fascicle length were determined for each muscle; these length changes were utilized in subsequent predictions of muscle force (i.e. active and passive contributions). Contraction velocity calculations were not required because all of the test conditions were isometric.

2.3.4.3 Calculation of Moments

In order to calculate the external demand, scaled digital photographs were taken for each posture. Using a symmetrical, 2D biomechanical model (Q-Back, Queen's University), subjects' postures were digitized (hand, wrist, elbow, shoulder, ear canal, C7/T1, and L4/L5) and input into the software model, accompanied by subjects' anthropometric data, to calculate the net L4/L5 extensor moment required to maintain static equilibrium. To minimize error, each trial was digitized three times and the outputs were subsequently averaged. Following this step, the net muscle moment required for equilibrium was calculated as the residual moment that remained once the restorative moment provided by passive structures had been accounted for; namely, that provided by the L4/L5 disc and the PLAD. The passive moment in the L4/L5 disc was calculated based on the intervertebral rotations predicted by the model, using the equations presented in McGill (1988).

PLAD extensor moments were estimated using data obtained from the digital photograph and force measurements from the in-series load cell. From each 2D photograph, the PLAD force unit vector $(\vec{u})$ and the radius, or muscle line of action vector $(\vec{r})$ were determined using custom software (Q-Dig, Queen's University). The vectors were calculated through digitization of the L4/L5 joint as well as the PLAD element line of action (top of the shoulder to the dorsal projection of the PLAD waist belt). The effective moment arm or unit moment vector of the PLAD about the flexion/extension axis (r_z) was determined from the cross product of these vectors. The extensor moment created by the PLAD was then calculated as the product of PLAD force (F_{PL}) and moment arm (r_z). The PLAD reaction force vector $(\vec{F}_{PL})$ was also determined by multiplying the PLAD force (F_{PL}) measured from the load cell by the PLAD force unit vector $(\vec{u})$. To minimize error from digitization, this process was repeated three times and all results were subsequently averaged.

With the contributions of the passive structures known, the net flexion/extension moment required by the musculature was calculated using:

$$\sum_{m=1}^{58} M_{mz} = M_{L4/L5z} - M_{dz} - M_{PLz} \qquad [2]$$

Where M_{mz} = muscle moment (Nm) about the z (flexion/extension) axis, $M_{L4/L5z}$ = reaction moment required for static equilibrium (Nm) about the z axis, M_{dz} = flexion/extension moment provided by the disc (Nm), and M_{PLz} = Flexion/extension moment provided by the PLAD (Nm).

2.3.4.4 Calculation of Muscle Forces and Moments

Raw EMG data were processed and reduced by calculating the root mean square (RMS) amplitude of each muscle signal. Following the removal of baseline noise, EMG amplitude data from the MVC trials were treated with a 150 ms sliding window; after processing each of the repeated MVC trials, the largest RMS amplitude found within a given window was used to represent 100% MVC. These values represented the maximal activation of the muscle, and were used to normalize trial data with respect to maximum voluntary output. For the experimental trial data, baseline noise was again removed, and the RMS amplitude of the EMG signal recorded over the trial was used to represent the activation state of each muscle. EMG signals were then normalized with respect to maximum and expressed in %MVC. Muscle force predictions for the RA, EO, IO, TES, LES, MULT, QL, and PS fascicles were then determined based on these normalized signals. Activation states for QL and PS muscles were predicted using the LES and IO signals, respectively, as suggested by Cholewicki and McGill (1996).

For each fascicle (n = 58) total muscle force was determined using the following equation:

$$F_m = Gain[EMG_m * PCSA_m * \delta * L_m + F_p] \quad [3]$$

Where F_m = muscle force (N), EMG_m = normalized EMG level (%MVC), $PCSA_m$ = muscle physiological cross-sectional area (cm^2), δ = maximum muscle stress, set at 35 N/cm^2 (Brown and Potvin, 2007), L_m = muscle force/length coefficient determined from anatomical model and equations presented in McGill (1986), F_P = passive elastic force in the muscle, determined from anatomical model outputs and equations presented in McGill (1986), $Gain$ = multiplier used equate the EMG-determined muscle moment and the residual moment.

The EMG-predicted muscle forces for the 58 fascicles were combined with their skeletal geometry to yield force vectors $(\vec{F}_{m_i})$ describing each muscle fascicle. The predicted moments generated by each fascicle were then summed to yield an estimate of net muscle moment. Finally, a common "Gain" factor was applied to the muscle force predictions in order to balance the EMG-based muscle moment with the moment required for equilibrium, as predicted from the linked-segment biomechanical analysis (Equation 3).

2.3.4.5 Calculation of L4/L5 Joint Compression and Shear Forces

Following application of the gain factor and subsequent moment balance, the predicted muscle forces $(\vec{F}_m)$ obtained from the model were summed to produce a net muscle force vector $(\vec{F}_{Mnet})$ acting about L4/L5. The F_z component of the vector was ignored, as this value was always zero because the tasks were sagittal symmetric, causing no net force or moment about the lateral bend axis. Therefore, only the F_x and F_y components of muscle force were retained and combined with the reaction forces of both the upper body mass $(\vec{F}_{BW})$, and the PLAD $(\vec{F}_{PL})$ to produce a 2D joint force vector $(\vec{F}_J)$ that described the net joint load acting at L4/L5 (in global co-ordinates):

$$\vec{F}_J = \vec{F}_{Mnet} + \vec{F}_{BW} + \vec{F}_{PL} \quad [4]$$

Where $\vec{F}_J$ = force acting on the L4/L5 intervertebral joint (N), $\vec{F}_{Mnet}$ = net muscle forces acting on the joint (N), $\vec{F}_{BW}$ = reaction force due to the body weight above L4/L5 (N) (from Winter (1990), $\vec{F}_{PL}$ = PLAD reaction force (N).

Intervertebral joint compression and shear forces were quantified through projection of the joint force vector onto the local axes of the L4/L5 joint. The orientation of the L4/L5 joint (relative to global co-ordinates) was determined for each posture based on the data of White and Panjabi (1990):

$$\alpha_{L4/L5} = (\theta_L * L * \psi_{L5}) + b \quad [5]$$

Where $\alpha_{L4/L5}$ = rotation that describes the compression and shear axes of L4/L5 relative to the global co-ordinate system, θ_T = trunk flexion (rads), L = proportion of trunk flexion occurring in the lumbar spine (set to 0.513 based on data from White and Panjabi (1990)), $\psi_{(L5)}$ = %flexion assigned to the L5 vertebra (23.6%), b = orientation offset of the joint in neutral posture = 0.209 rads (~ 12 degrees) from White and Panjabi (1990).

With the rotation angle known, the net joint forces ($\vec{F}_J$) were projected onto the local vertebral axis system to yield L4/L5 compression and shear forces through formulation of the rotation matrix [R]:

$$[R][\vec{F}_J] \Rightarrow \begin{bmatrix} \cos\alpha_{L4/L5} & -\sin\alpha_{L4/L5} \\ \sin\alpha_{L4/L5} & \cos\alpha_{L4/L5} \end{bmatrix} \begin{bmatrix} F_x \\ F_y \end{bmatrix} = \begin{bmatrix} F_s \\ F_c \end{bmatrix} \qquad [6]$$

Where $\vec{F}_J$ = the global joint force vector, $\alpha_{L4/L5}$ = the angle describing the orientation of the intervertebral axes relative to the global frame, F_S = A-P shear force at L4/L5 (N), F_C = compressive force at L4/L5 (N).

2.3.5 Statistics

2.3.5.1 Descriptive Statistics

For each experimental condition, descriptive statistics were tabulated to quantify the mechanics of the PLAD system (forces, moment arm, moments, and changes in EMG activity). The purposes of these data were two-fold. First, these data were quantified to provide a basis of comparison to previous research involving the device. Additionally, these data were collected for use in the development of regression equations that could be used during implementation of the device to pre-configure the system and thus provide a specific amount of assistance.

2.3.5.2 Repeated Measures ANOVA

Compression and shear forces were quantified for all subjects within each of the 30 experimental conditions. The effect of the PLAD on L4/L5 compression and shear joint loading was assessed using a two-factor (Angle x Stiffness), 5 x 6 repeated measures ANOVA. For the ANOVA tests, significant main effects and interactions (α <0.05) were contrasted post-hoc using paired t-tests and a procedure known as false detection rate (FDR). As suggested by Benjamini & Hockberg (1995), FDR is a method better suited for multiple comparisons when contrasted with other overly conservative techniques, such as the Bonferroni correction procedure. In brief, the FDR technique attains a higher level of power by controlling the proportion of errors that exist among only the tests that had a null hypothesis rejection. All statistical analyses were performed using SPSS 11.0 for Windows (SPSS Corporation, Chicago, IL). Trial observations were also described in terms of relative and absolute PLAD effectiveness.

2.3.5.3 Regression Equations

Using study results, regression equations were developed to predict the sagittal moment arm of the device as well as the accompanying reduction in joint compression that would occur throughout a range of trunk motion and PLAD assistive force. The moment arm prediction model was developed for future use and/or prescription of the PLAD device. As such, subject anthropometry was considered, in order to account for variability in the size of the effective moment arm of the PLAD (across a range of trunk motion) that is due, in part, to the subject's/user's trunk depth and torso length. The regression model predicting changes in joint loading utilized compression forces calculated from the EMG-based model for each of the experimental conditions. Because PLAD force varied within posture, it was hypothesized that an equation could be developed that would predict the benefits of wearing the PLAD for postures involving varying amounts of trunk flexion, and at the same time, account for any co-activation of the trunk flexors in response to resistance of the PLAD system.

2.4 Results

2.4.1 PLAD Forces, Moments and EMG Data

Forces produced by the elements of the PLAD device were quantified and are presented in Figure 2-3, contrasted by trunk flexion angle and element stiffness. PLAD forces increased as a function of trunk posture, and for the most part, with use of the stiffer elastic elements, producing average (SD) forces of 39.18 (15.64) N, 86.61 (27.76) N, 140.91 (34.36) N, 187.64 (40.84) N, and 239.15 (50.23) N across each respective level of trunk flexion. The lack of a consistent linear increase in PLAD force with higher stiffness was most likely due to issues of fit which led to slipping at the shoulder and knee attachment points; this would have decreased the amount of strain (stretch) in the stiffer PLAD elements for a given amount of trunk flexion.

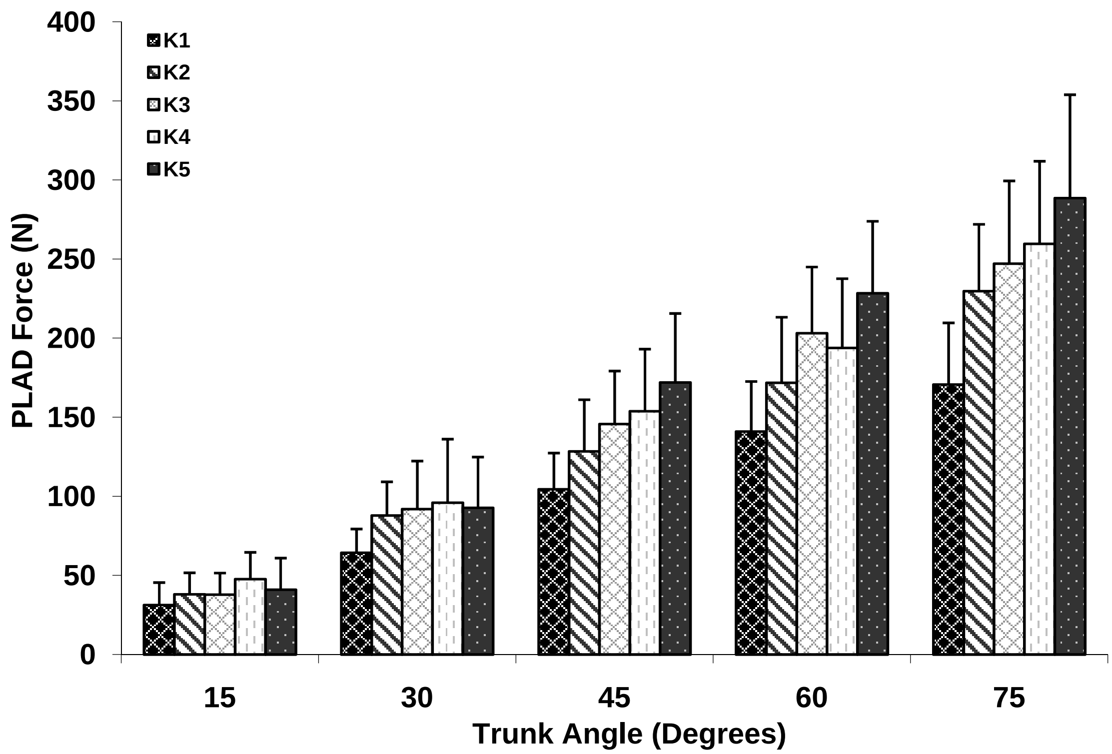


Figure 2-3 Average (SD) force (N) generated by the PLAD elements across trunk flexion conditions.

The sagittal moment arm (r_z) of the PLAD decreased with trunk flexion (Figure 2-4). Across postural conditions, the moment arm of the device decreased ~30% from 0.184 (0.012) m at 15 degrees of trunk flexion to a final value of 0.132 (0.02) m at 75 degrees of trunk flexion. Upon observation of Figure 2-4, it is also apparent that with increasing trunk flexion, the variability (SD) of the PLAD moment arm increased as well (by nearly 66% across subjects), reflective of fitting issues and anthropometric differences between subjects.

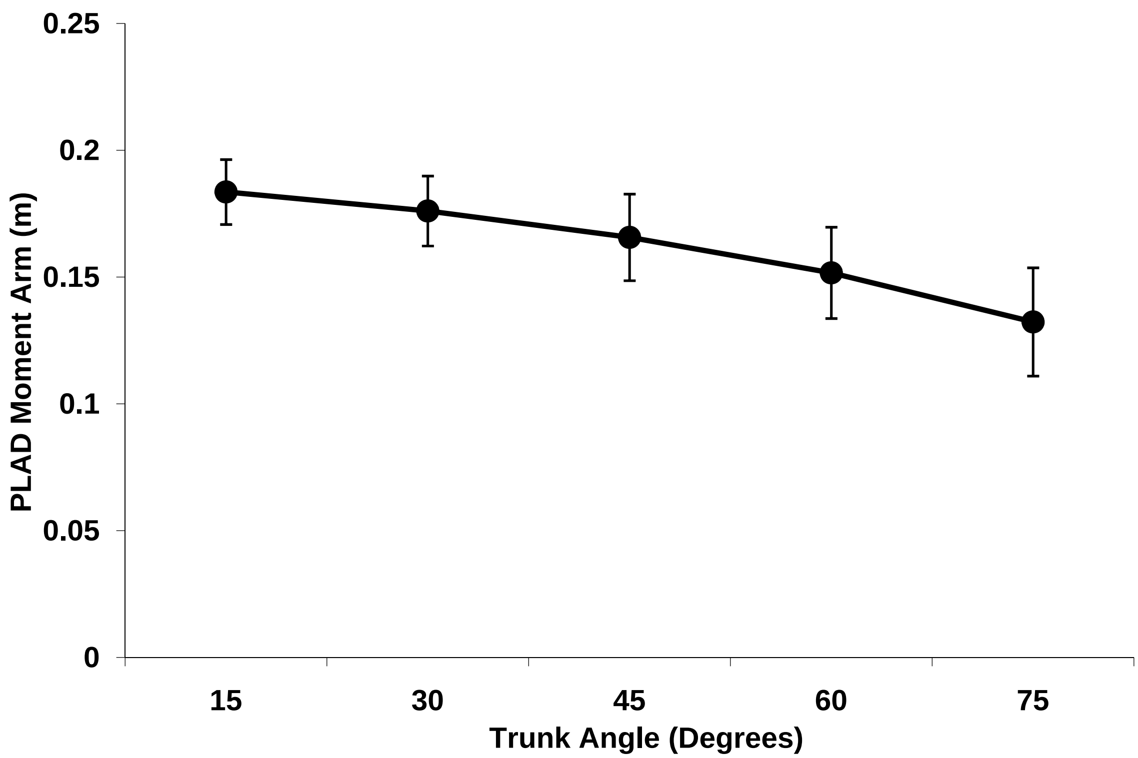


Figure 2-4 Change in PLAD Flexion/Extension moment arm (m) as trunk flexion increased. Error bars represent +/- 1 SD.

Accounting for the passive moment created by the PLAD, net (residual) L4/L5 moments were tabulated (Figure 2-5). From these data, the absolute and relative assistance provided by the PLAD were quantified. As expected, the amount of assistive moment generated by the PLAD increased as a function of trunk posture and PLAD element stiffness. Collapsed across stiffness, the PLAD produced: 7.28 (2.74) Nm, 15.14 (4.56) Nm, 23.19 (4.82) Nm, 28.74 (5.53) Nm and 31.99 (6.51) Nm of assistive moment at each respective level of trunk flexion. When expressed with respect to the No-PLAD condition, the PLAD provided 16.39 (7.2) %, 20.05 (8.55) %, 22.72 (6.74) %, 23.92 (5.32) %, and 23.84 (5.41) % of the total moment required within each postural condition, respectively (Table 1).

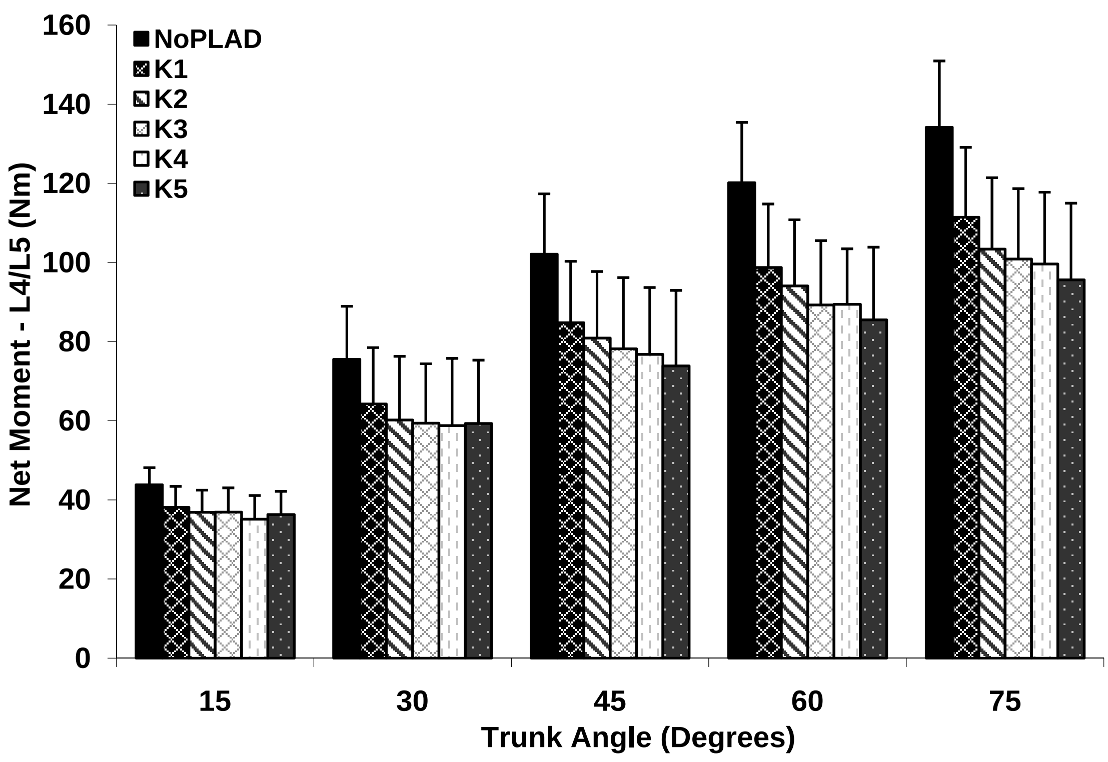
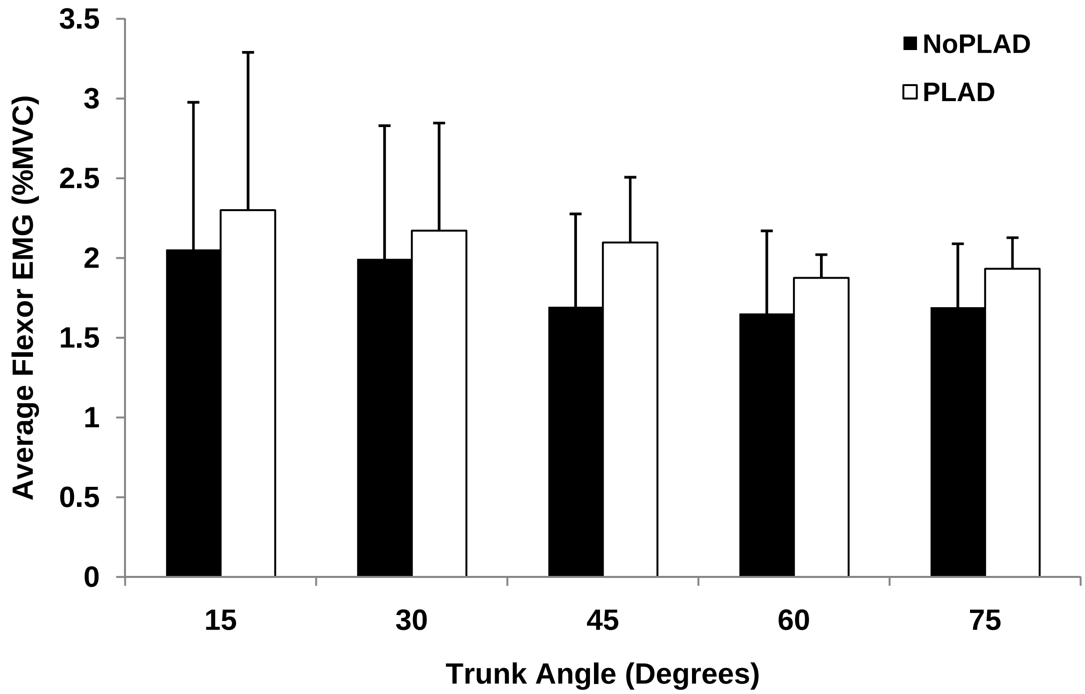


Figure 2-5 Average (SD) L4/L5 moment (Nm) generated by the PLAD contrasted by levels of element stiffness and trunk flexion.

Table 2-1 Relative level (%) of required trunk extensor moment provided by the PLAD contrasted across levels of element stiffness and trunk flexion.

PLAD Stiffness	Trunk Flexion Angle (Degrees)				
	15	**30**	**45**	**60**	**75**
K1	13.10 (6.44)	14.94 (5.19)	16.95 (4.64)	17.83 (4.37)	16.95 (3.98)
K2	15.91 (6.30)	20.27 (7.84)	20.75 (6.26)	21.71 (5.66)	22.94 (4.92)
K3	15.78 (6.74)	21.38 (8.75)	23.40 (6.95)	25.69 (6.69)	24.82 (6.12)
K4	19.91 (7.73)	22.18 (11.19)	24.80 (7.26)	25.56 (4.16)	25.74 (4.95)
K5	17.23 (8.75)	21.48 (9.81)	27.66 (8.62)	28.82 (6.76)	28.74 (7.10)
Mean (SD)	16.39 (7.19)	20.05 (8.55)	22.71 (6.74)	23.92 (5.32)	23.84 (5.41)

Absolute changes in observed EMG activity are presented in Figure 2-6. To illustrate the effect of the PLAD, EMG data were pooled across PLAD stiffness conditions and averaged to reflect activation changes due to use of the PLAD within the trunk flexor (RA, IO, and EO) and extensor (MULT, LES, TES, and LD) musculature. As shown in Figure 2-6, decreases in extensor EMG were paired with small increases in trunk flexor EMG, reflective of a mild increase in trunk muscle co-activation and a need for greater activation of the trunk flexors in order to oppose the resistance of the PLAD's elastic elements. Within the flexor musculature, the EO was observed to be the muscle most responsive to use of the PLAD, while the RA was the least responsive, with relative compensatory increases ranging from 28-71% and 0.37-1.5% respectively across levels of trunk flexion. In the trunk extensors, the MULT was observed to be the muscle most responsive to use of PLAD, while the TES was the least responsive, with relative compensatory decreases ranging from 4-62% and 16-38% respectively across levels of trunk flexion. It should be noted that flexion-relaxation was evident in a few subjects' data at 75 degrees of trunk flexion.



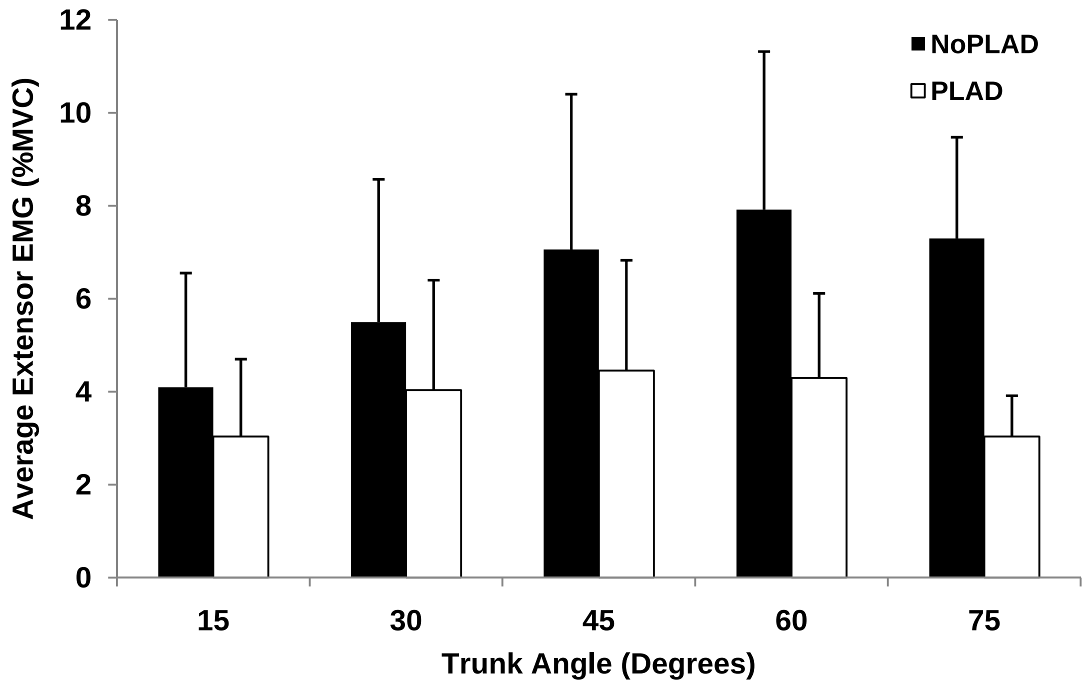


Figure 2-6. Top: Average (SD) increases in trunk flexor EMG (%MVC) due to use of the PLAD. **Bottom:** Average (SD) decreased in trunk extensor EMG due to use of the PLAD.

2.4.2 Effect of the PLAD on L4/L5 Compression

Compressive joint loads obtained from the EMG model are presented in Figure 2-7. ANOVA diagnostics revealed that for the repeated factor of Angle, and the Stiffness x Angle interaction, assumptions of sphericity were violated (based on Mauchly's W statistic). To correct the violation, the associated degrees of freedom (df) were adjusted using the Greenhouse-Geisser method. Once corrected, the repeated measures ANOVA indicated significant main effects for Angle, $\underline{F}$ (1.35, 10.8) = 29.54, p = 0.0001 and Stiffness, $\underline{F}$ (2.65, 21.25) = 12.216, p = 0.0001, reflective of an increase in biomechanical demand with greater trunk angle and an increase in PLAD effectiveness through use of increasingly stiffer elastic elements. However, a significant interaction was also identified between the two variables $\underline{F}$ (4.15, 33.22) = 3.076, p = 0.028. The interaction is visible within Figure 2-5; with increasing trunk flexion and PLAD element stiffness, greater reductions in compressive force were observed.

Based on the FDR algorithm and an a priori α level = 0.05, post hoc contrasts were deemed significant when $p \leq 0.008$. Using this criterion, paired comparisons were performed within each level of angle to identify instances where use of the PLAD significantly reduced the estimated joint compression acting at L4/L5. Flexed 15 degrees, the PLAD did not significantly reduce the compressive force, regardless of the level of PLAD stiffness. When flexed 30 degrees, a significant reduction in joint compression occurred in the trials using the stiffest PLAD element (K5) (p = 0.001). This equated to a 204.02 (113.45) N or 11% reduction in joint compression. In the 45 degree posture, 4 of the 5 PLAD configurations significantly reduced the net compressive load on the joint, by magnitudes ranging from 191.3 (121.9) N to 246.68 (149.41) N (K1, 2, 4, and 5; p = 0.0001, p = 0.002, p = 0.001, and p = 0.0001), corresponding to a decrease in joint compression that ranged from 8-12%. At 60 degrees of trunk flexion, the PLAD significantly reduced joint compression by approximately 14% when the K4 and K5 elements were used

(369.75 (340.33) N, $p = 0.007$, and 373.46 (292.65) N, $p = 0.004$)). When subjects flexed to 75 degrees, net joint compression was significantly reduced with 4 of the 5 PLAD elements, with effects ranging from 447.42 (314.16) N to 646.12 (404.71) N (K1, 3, 4, 5; $p = 0.004$, $p = 0.003$, $p = 0.001$, $p = 0.001$, respectively). In relative terms, this equated to a 10-13% decrease in joint compressive load. Relative changes in joint compression due to use of the PLAD are summarized in Table 2-2.

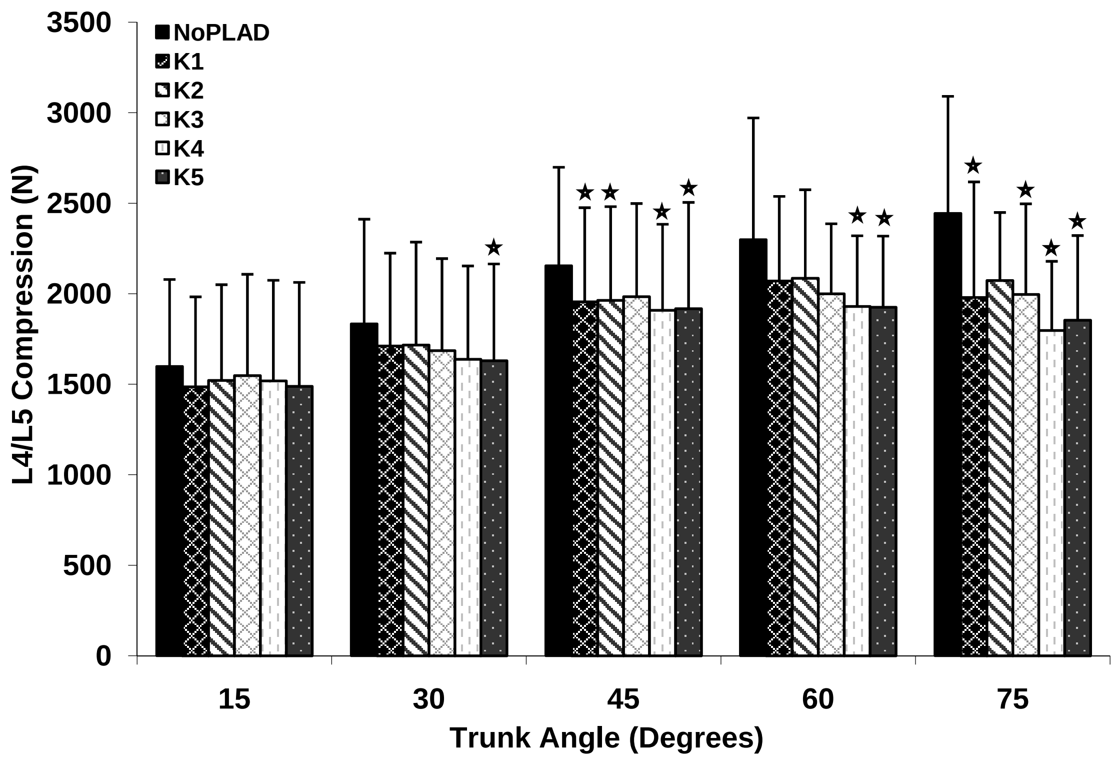


Figure 2-7 Effect of the PLAD on L4/L5 joint compression, as estimated by the EMG-based model. Average (SD) compression values are presented. Significant differences between PLAD and No-PLAD conditions are indicated by ☆.

Table 2-2 Relative change (%) in L4/L5 joint compression (N) contrasted by PLAD element stiffness and levels of trunk flexion. Bold, italicized values correspond to trials where significant decreases were observed (p < 0.018).

PLAD Stiffness	Trunk Flexion Angle (Degrees)				
	15	**30**	**45**	**60**	**75**
K1	7.23 (8.97)	5.59 (13.70)	***9.45*** ***(3.52)***	8.10 (9.20)	***19.08*** ***(12.67)***
K2	5.21 (13.69)	6.26 (9.26)	***8.93*** ***(5.73)***	7.86 (7.02)	13.21 (10.84)
K3	4.33 (7.22)	7.34 (10.58)	7.77 (8.27)	10.83 (9.41)	***17.70*** ***(10.85)***
K4	5.57 (14.25)	9.74 (14.89)	***11.12*** ***(6.58)***	***13.78*** ***(10.64)***	***24.77*** ***(13.06)***
K5	8.39 (10.46)	***11.28*** ***(5.95)***	***11.99*** ***(7.05)***	***14.29*** ***(8.01)***	***23.30*** ***(10.45)***
Mean (SD)	6.14 (10.91)	8.04 (10.88)	9.85 (6.23)	10.97 (8.85)	19.61 (11.57)

2.4.3 Effect of the PLAD on L4/L5 Shear

As with the compression data, assumptions of sphericity could not be met, requiring the Greenhouse-Geisser correction method to be applied to all main effects and interactions. The results of the repeated measures ANOVA indicated a significant main effect for Angle, $\underline{F}$ (1.72, 13.77) = 94.245, p = 0.0001, as well as a significant interaction $\underline{F}$ (3.41, 27.26) = 3.296, p = 0.031, suggesting significant changes in shear force with increases in trunk angle and PLAD stiffness. When analyzed post hoc, it was determined that the shear forces acting on the joint changed significantly as trunk angle increased from 15 to 75 degrees (Figure 2-8). Based on the FDR algorithm and an a priori α level = 0.05, post hoc contrasts were deemed significant when $p \leq 0.028$. Using this criterion, paired comparisons were performed within each level of angle to identify instances where use of the PLAD significantly altered the estimated amount of joint shear acting at L4/L5. From 15 to 60 degrees of trunk flexion, no significant effects were identified. When subjects flexed to 75 degrees, net posterior joint shear was significantly increased with use of 4 of the 5 PLAD elements, with effects ranging from 72.79 (113.91) N to 92.57 (92.03) N (K2, 3, 4, 5; $p = 0.017$, $p = 0.005$, $p = 0.02$, $p = 0.026$, respectively). In relative terms, this equated to a 110-140% increase in posterior joint shear. Relative changes in joint shear due to use of the PLAD are summarized in Table 2-3. While minimal significant differences were found, the results are functionally relevant. The reaction force created by the PLAD system created a marked difference in shear loading across all flexion conditions; however, this was quite variable and thus hindered tests of significance. This variability is due, in part, to the sensitivity of the EMG-based model as well as the effectiveness of the PLAD, and it will be addressed in the discussion.

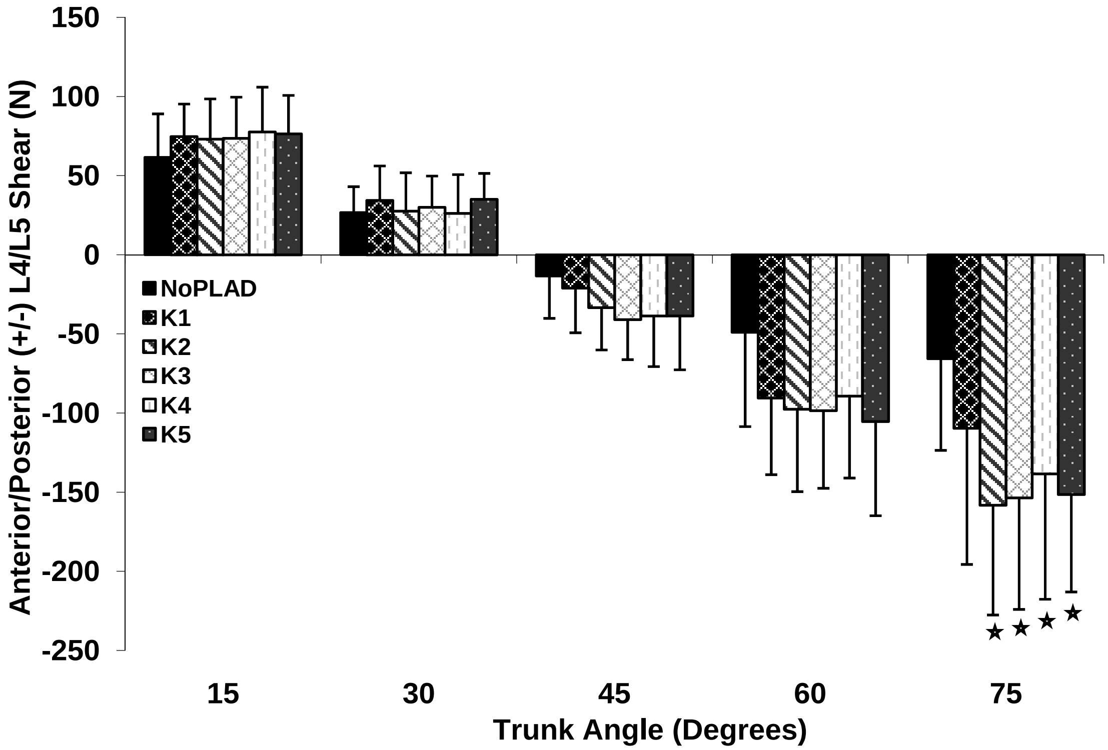


Figure 2-8 Effect of the PLAD on L4/L5 joint shear, as estimated by the EMG-based model. Average (SD) L4/L5 shear forces are presented. Significant differences between PLAD and No-PLAD conditions are indicated by ☆.

Table 2-3 Relative change (%) in L4/L5 joint shear (N) contrasted by PLAD element stiffness and levels of trunk flexion. Bold, italicized values reflect trials where significant decreases were observed (p < 0.028).

PLAD Stiffness	Trunk Flexion Angle (Degrees)				
	Anterior Shear		Posterior Shear		
	15	30	45	60	75
K1	21.37 (26.67)	28.63 (94.17)	56.10 (92.76)	84.45 (91.62)	66.88 (30.92)
K2	18.74 (45.68)	3.316 (63.44)	147.67 (36.06)	98.97 (94.57)	***140.95*** ***(49.14)***
K3	19.65 (24.51)	12.47 (81.97)	203.68 (53.43)	100.83 (41.46)	***133.85*** ***(47.75)***
K4	26.08 (38.29)	-2.12 (83.70)	186.95 (39.64)	82.12 (57.49)	***110.84*** ***(64.89)***
K5	24.18 (23.64)	31.35 (44.48)	187.04 (41.74)	114.84 (49.41)	***130.64*** ***(58.60)***
Mean (SD)	22.01 (31.76)	14.73 (45.47)	156.291 (52.85)	96.25 (66.91)	116.64 (50.26)

2.4.4 Predictive Equations

2.4.4.1 PLAD Moment Arm

Trial and anthropometric data from every PLAD condition (n = 225) were used within a step-wise regression analysis to develop a predictive equation for estimation of the PLAD sagittal moment arm across a range trunk flexion. Trunk angle, subject height and subject weight were found to be significant predictors ($p < 0.05$) of the sagittal moment arm of the PLAD. Combined, the three variables accounted for 90.1 % of the variance in moment arm across subjects and postures with a standard error (S.E.) of 0.007 m (Figure 2-9).

From the regression analysis, the following equation was obtained:

$$r_z = -0.195 + 0.192(HT) + 0.001(BW) - 0.001(TA) \quad [7]$$

Where r_z = sagittal moment arm of the PLAD device (m), HT = Height (m), BW = Body weight (kg), and TA = Trunk flexion, or angle, with respect to vertical (deg).

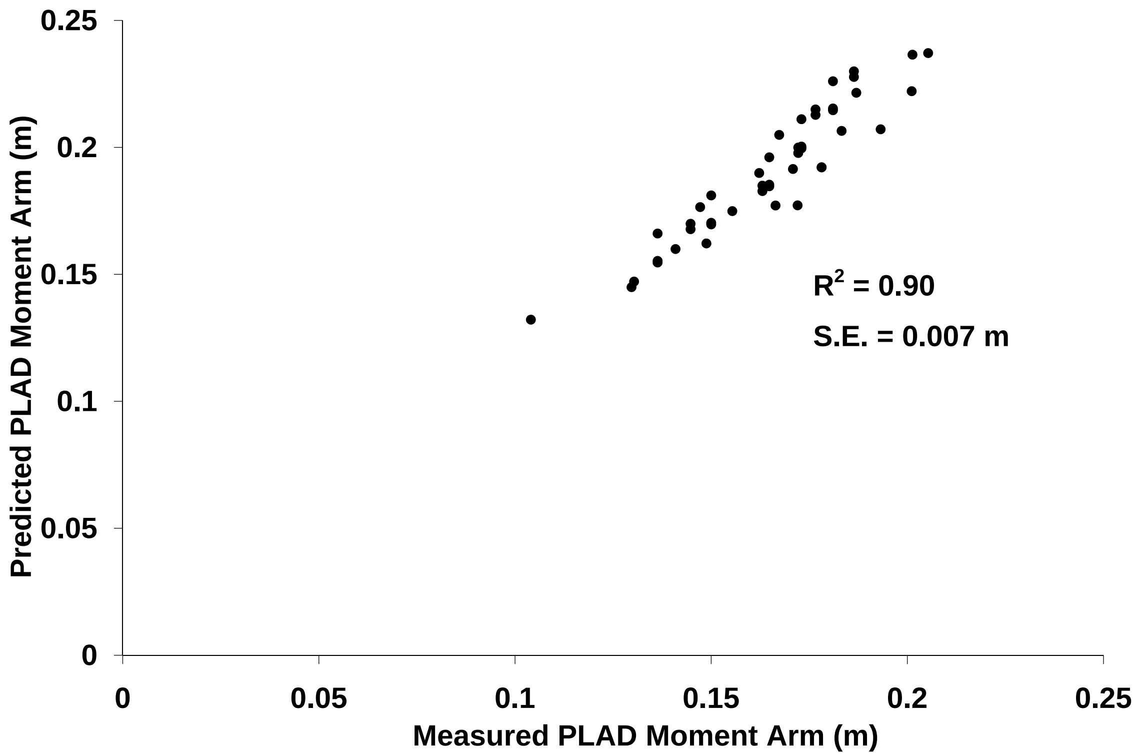


Figure 2-9 X/Y Scatterplot depicting agreement between actual PLAD moment arm (m) and predicted values obtained using the regression equation. **Note**: The PLAD's moment arm about L4/L5 decreases through trunk flexion and moves from the top right to the bottom left of the plot.

2.4.4.2 Joint Compression

As with the moment arm analysis, all trial data (n = 270) were entered into a step-wise regression analysis to predict the expected joint load (compression) that would result from use of PLAD. Body weight, PLAD force, and the sine of the trunk angle were all found to be significant predictors ($p < 0.05$) of L4/L5 joint compression. Together, these variables accounted for 70 % of the observed variance in joint compression, with a standard error (S.E.) of 300.45 N (Figure 2-10). The regression analysis produced the following equation:

$$F_c = -26.22350 + 47.362(BW) + 1158.089(\sin TA) - 1.677(F_{PL}) \qquad [8]$$

Where F_c = Net L4/L5 joint compression (N), BW = Body weight (kg), TA = Trunk flexion or angle, with respect to vertical (deg), and F_{PL} = Total assistive force provided by the PLAD system (N)

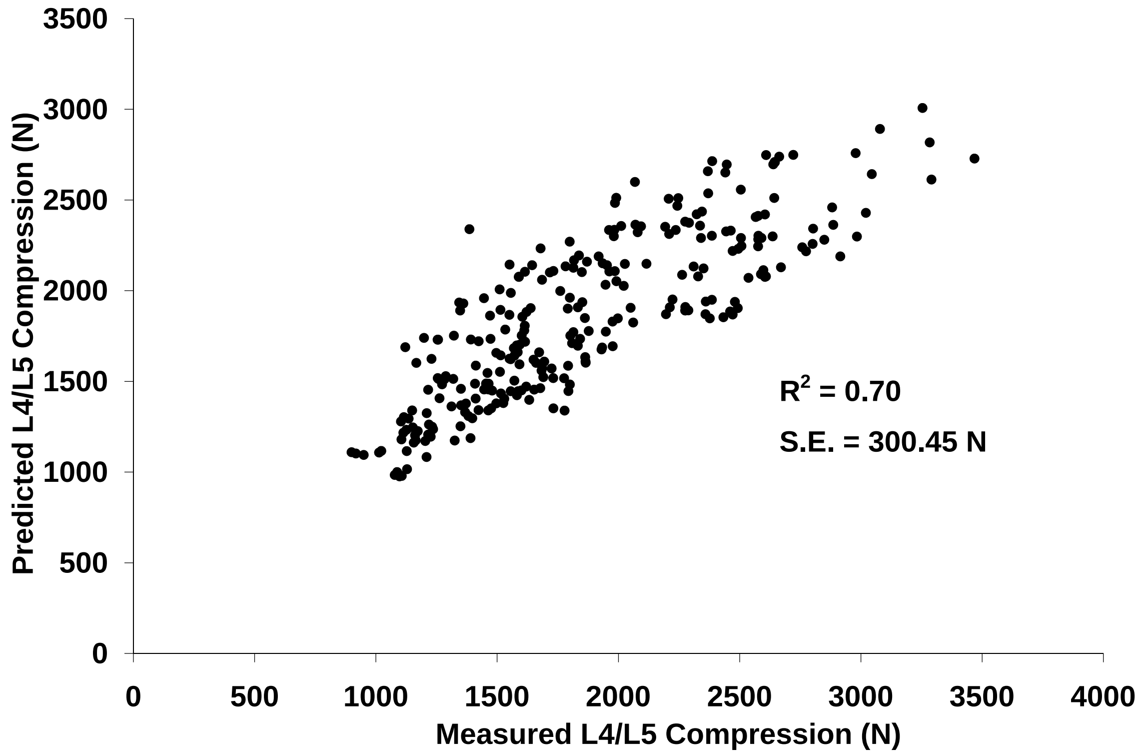


Figure 2-10 X/Y Scatterplot depicting agreement between L4/L5 joint compression values (N) and predicted values obtained using the regression equation.

2.5 Discussion

The objective of this study was to assess the effect of the PLAD system at reducing spinal loads at the L4/L5 joint. To do so, changes in spinal load attributable to use of the PLAD were documented across a range of flexed trunk postures and levels of PLAD assistive support. This provided a comprehensive assessment of the mechanics of the PLAD, how these changed as a function of trunk posture, and ultimately, the resulting effects experienced at the L4/L5 joint. Relative activation states (EMG) recorded from select trunk flexor and extensor muscles were quantified and input into an EMG-assisted biomechanical model. These data were combined with reaction forces and moments produced by the PLAD and upper body mass to predict net compression and shear forces at the L4/L5 joint. In general, the results of the study support previous research by Abdoli et al. (2007), who suggested that use of the PLAD with static flexed postures and/or lifting tasks will result in a net decrease in the net joint loads incurred at L4/L5.

As expected, model-based predictions of joint compression increased with trunk flexion. In terms of magnitude, the values predicted via the EMG-based model are consistent with what has been reported in previous research involving use of this model, its associated predictive equations, and muscle lines of action (Potvin et al., 1991; Brown & Potvin, 2007). With increasing trunk flexion, significant reductions in joint compression were observed due to use of the PLAD. This was documented across a range of stiffness levels, and as such, increasing levels of assistive support (i.e., force and moment). As a general observation, due in part to the variability in results associated with experimental error and anthropometric differences between subjects, relative decreases of 9% or more were required for significant differences to be identified between PLAD and No-PLAD condition (Table 2-2). In absolute terms, this equated to a quantifiable reduction in net compressive force that ranged from 191 to 646 N; across conditions where the trunk was flexed 30 to 75 degrees.

Abdoli et al. (2007), mathematically expressed the effect of the PLAD on L4/L5 compression to be directly related to the mechanical advantage of the PLAD elements over the erector spinae. In their work, the erector spinae were represented as a single equivalent muscle vector with a moment arm of 0.06 m. Accordingly, the absolute effect of the PLAD at reducing L4/L5 joint compression can be estimated to be $\downarrow F_c = F_{PL}[(\text{PLAD moment arm}/0.06) - 1]$. Based on this relationship, using average PLAD force (Figure 2-3) and moment arm data (Figure 2-4) quantified in the current study, we would expect to find average compressive force decreases of approximately: 80, 167, 248, 286, and 288 N across each respective level of trunk flexion. Using the EMG-based modeling approach, our results were similar, as average estimates of PLAD effectiveness were quantified to be: 86, 157, 209, 297, and 503 N across trunk flexion levels. This implies that the small observable increases in co-contraction observed across conditions (Figure 2-6) did not counteract the benefits of the PLAD in terms of reducing joint compressive load. A large difference is evident when the hypothetical and observed results taken from the 75 degree condition are compared; this is most likely due to significant changes in the moment-generating capacity of the erector spinae and multifidus at lumbar end range of motion. These effects were accounted for in the EMG model; however, the erector spinae moment arm used in the Abdoli et al. (2007) mathematical proof is a constant and is insensitive to the effects of trunk flexion.

The PLAD was found to have significant effects on the net joint shear acting at L4/L5. Towards joint end range of motion, use of the PLAD significantly increased posterior joint shear at L4/L5 by magnitudes ranging 70 – 90 N (110-140%). Across all levels of trunk flexion, with and without use of the PLAD, shear force estimates were found to be quite variable, especially in comparison to the compressive force data. The inherent sensitivity of shear force predictions through EMG-based models has been noted in the literature (Nussbaum et al., 1995). In the current study, this is most likely attributable to the low activation levels associated with maintaining the postures, combined with the geometry of the EMG-based model. However, the results obtained in the No-PLAD condition are comparable to previous research that has used this model within studies involving low activation levels and moderate levels of trunk flexion (Potvin et al., 1991; Callaghan et al., 1999). In comparison to the theoretical proof by Abdoli et al. (2007), there is very little agreement with reference to the results obtained in the current study in terms of the effect of the PLAD on joint shear. While considerable variability was observed, the results do suggest that the PLAD has a significantly greater effect on L4/L5 joint shear than previously reported. This can be explained due to differences between modeling strategies. In

Abdoli et al. (2007), a single rigid body was used to represent the trunk, and as such, trunk rotation was assumed to take place entirely about one joint. As such, this rotation angle describes the orientation of the local compression and shear axes of the vertebrae, and is also used in subsequent projection of the PLAD reaction force vector onto these axes. In the current study, as the trunk flexed forward, the EMG model proportionately distributed motion across the lumbar vertebrae based on observations of Panjabi and White (1990). As such, the local L4/L5 compression and shear axes did not deviate (rotate) to the same extent as described in Abdoli et al. (2007). As trunk flexion increased, the PLAD reaction force became more aligned with the shear axis of L4/L5 and as such, created assistive, posterior-directed shear forces much larger than those suggested in Abdoli et al. (2007). Based on these findings, it is apparent that the PLAD has a more positive effect at reducing anterior shear forces about L4/L5 than has been previously documented. This is due to the line of action of the PLAD reaction force relative to the orientation of the L4/L5 shear axis, particularly towards end range of trunk motion.

If the PLAD is to be recommended as a form of ergonomic control, it is critical that its effectiveness is quantifiable, repeatable, and adjustable (scalable), relative to the physical demands of a task and the anthropometry of the prospective user. As such, a secondary objective of the current study attempted to develop regression equations that could be used to prescribe or determine a given amount of assistive support based on task constraints and user anthropometry. In the development of the moment arm regression equation, body height, weight, and flexion angle were all found to be significant predictors of the moment generating capacity of the PLAD. Therefore, for a given posture, the effective moment arm of the PLAD is different depending on the user. This was evident in the current study, and it is reflected in the larger variability in moment arm estimates that developed as trunk flexion increased (Figure 2-4). Factors contributing to this include trunk length (from the pelvis to the shoulder) and torso depth. With this in mind, stature and body weight were assumed to be correlates of these measures and used as inputs into the regression model. Using this approach, the regression model fit the data quite well and explained a large percent of the observed variance (R^2 = 0.90, S.E. = 0.007 m). Using a similar approach, a second regression equation was developed to predict changes in L4/L5 compression due to use of the PLAD. Body weight, PLAD force, and trunk angle were found to be significant predictors of low back compressive load, explaining approximately 70% (S.E. = 300 N) of the variance associated with the data set.

It should be noted that these equations are exploratory in nature and do not extend past the results of the current study. They were examined to test the notion that PLAD effectiveness could be predicted based on postural data and a user's stature and weight. Based on the relative strength of the developed equations and the limited data set used to create them, the results are encouraging. They serve as preliminary results that illustrate the novelty of "sizing" or predictive equations that can be used in future applications to configure the PLAD system to a predetermined amount of support, dependent on work posture and user anthropometrics.

The results of this study further illustrate the potential benefits associated with use of the PLAD as a means of reducing low back loads incurred during static flexed postures and lifting tasks. They are consistent with the theoretical proof developed by Abdoli et al. (2007) and other research that has documented the potential benefits associated with lifting belts (Granata et al., 1997; Kingma et al., 2006) and lumbosacral orthoses (Cholewicki et al., 2007). However, the results are suspect to several limitations that should be acknowledged. The research methodology used in the current study required subjects to flex forward to a predetermined static posture. Furthermore, subjects were not required to hold any load in the hands. These steps were taken in order to minimize potential confounds due to postural differences naturally associated with use of the PLAD and differences in lifting dynamics, as well as any effects of increased co-contraction due to an external load in the hands. As the PLAD is a purely passive system, the data presented are thought to solely represent the benefits associated with the biomechanical advantage of the device across a range of trunk flexion. Future research should attempt to identify additional positive/negative attributes associated with use of the PLAD during dynamic, repetitive tasks and determine how these attributes affect compressive and shear forces acting on L4/L5. Furthermore, the current study focused on the effect of the PLAD on L4/L5 joint loads during symmetrical flexed postures. Additional research with an EMG-based model is encouraged, and researchers should attempt to document the effectiveness of the PLAD at reducing spine loads associated with three-dimensional asymmetrical lifting tasks, by undertaking an analysis that spans the entire lumbar spine, as opposed to one intervertebral level. In addition, while earlier research assessed shoulder and knee discomfort using a subjective ratings scale, a (Abdoli et al., 2006) future investigations should attempt to quantify the increased biomechanical loading that occurs at the shoulders and knee attachment points, and address whether or not use of PLAD actually reduces injury risk at one anatomical site (i.e., low back) and conversely increases injury risk at another (i.e., knee ligaments). Once these questions are assessed, the effectiveness of this system as an ergonomic control should be investigated using a longitudinal research study that quantifies peak and cumulative reductions associated with use of the PLAD and associated changes in reporting of low back pain and/or injury.

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Chapter 3
The Impact of a Personal Lift Assist Device (PLAD) on Lifting Kinematics and Co-ordination

3.1 Abstract

Workplace postures and trunk kinematics have been identified as significant risk factors for low back pain and/or injury. As such, ergonomic controls should attempt to minimize workplace exposures to these risk factors. Within our laboratory, we have been developing an on body lift assist device called the Personal Lift Assist Device (PLAD). This system is worn by workers and features elastic elements that stretch during trunk flexion to create an assistive moment about the trunk. Preliminary evidence has suggested that use of the device reduces erector spinae activity and minimizes localized muscle fatigue during repetitive lifting tasks. However, it is unknown what effect use of the PLAD may have in terms of lifting postures, trunk kinematics, and segmental co-ordination during repetitive lifting tasks. The purpose of the current study was to address this problem. Over two randomized testing sessions, 10 male subjects were required to complete a 15-minute repetitive lifting task with and without use of the PLAD. The task required lifting of a load (20% MVC) from floor to knuckle height at a rate of 6 lifts per minute. Subjects were instructed to lift with self-selected technique. Kinematic data describing motion about the lumbar spine, hip, and knee were used to quantify lifting posture, lifting velocities, and co-ordination via estimation of relative phase angles. The results of the study suggest that use of the PLAD causes users to lift with significantly less lumbar flexion and significantly greater flexion about the hip. Significant decreases were also observed in terms of the relative phase angle that existed between the lumbar-hip, and lumbar-knee segments, suggesting more synchronous motion occurred between these segments when the PLAD was worn. The results of the study further illustrate the potential of the PLAD as a viable ergonomic control solution; however, the differences observed in co-ordination-based measures reflect a

distinct change in lifting technique, and further research is needed to understand the long-term effects related to use of the PLAD.

3.2 Introduction

The financial and societal impacts of work-related musculoskeletal disorders (WMSDs) are well known (van Tulder, Koes & Bombardier, 2002; Luo et al., 2004; Punnet et al., 2005). In an effort to reduce the burden of WMSDs, researchers have attempted to establish causal links between injury onset and workplace risk. As a result, epidemiological research has indentified that personal traits, psychosocial factors and physical (biomechanical) exposures are significant contributors for the onset of WMSDs. While shown to pose independent risk, these factors are also known to interact positively, creating additive risk within certain workplaces (Marras et al., 1995). Amongst an extensive list of physical risk factors, manual lifting tasks, working postures (e.g., static lumbar flexion, stooped postures), and specific kinematic lifting variables (e.g. trunk flexion velocity) have been identified as significant work related attributes that contribute toward the development of low back injuries and/or pain (Norman et al, 1998; Marras et al, 1995).

With this in mind, preventative ergonomic control strategies attempt to reduce an operator’s exposure to such risk factors; intuitively, the most effective way is to completely design these potential risks out of the workplace, or assembly process. In a situation where workplace redesign is not possible, alternate control strategies are required; such strategies have included the implementation of training programs and the use of assistive devices like lifting belts and load transfer systems. While the scientific literature dedicated to the effectiveness of lifting belts has been deemed inconclusive (NIOSH, 1994), several researchers have been able to demonstrate that lifting belts have beneficial influence on lifting posture, lifting kinematics and lifting co-ordination; most notably they promote a decrease in lumbar flexion with a concomitant increase in hip flexion, as well as a decrease in lifting kinematics (i.e., flexion/extension velocity) (Granata et al, 1997; McGorry & Hsiang, 1999; Sparto et al., 1998). Among these researcher's studies, it was noted that results were varied by the style of lifting belt that was tested, which implies that lifting belts of one design are arguably better than another. For example, Granata et al. (1997) illustrated that an elasticized lifting belt that spanned the thorax and pelvis was more effective than both a stiff leather belt and an orthotic brace in terms of lifting kinematics and in ability to reduce spinal loads.

Furthermore, researchers have suggested that considerable motor learning must occur in order for a user to maximize any potential advantage associated with use of a material handling system, or device (Chaffin et al., 1999). For these reasons, it could be argued that part of the variability observed in previous research investigating terms effectiveness of ergonomic devices such as a lifting belts could be due to differences in product features/design and/or the level of participant training prior to the experiment. As such, adequate training or practice is required prior to any type of evaluation regarding use of an ergonomic aid, or assistive device.

For the last few years, research at our laboratory has focused on developing an ergonomic lifting aid (Personal Lift Assist Device – PLAD) designed to provide assistive support to the trunk extensor musculature during lifting tasks. To date, we have demonstrated its effectiveness at reducing erector spinae muscle demand in single and repetitive lifting tasks (Abdoli et al, 2006, 2007; Lotz et al., 2007). These results are comparable to those found with an elasticized lifting belt described in Granata et al. (1997). This makes sense, as the PLAD features elastic elements that connect the upper trunk to the pelvis in a similar manner to the elasticized corset used within that study. However, while significant changes in erector spinae amplitude have been documented in our previous work, research to date has focused on a minimal set of lifting tasks, and number of replications within a given experimental condition. Within this research (Abdoli et al., 2006) no appreciable changes were noted in terms of lifting kinematics at the lumbar spine and hip; however it must be recognized that the scope of these analyses was limited, as only a limited number of kinematic data were analyzed, and an analysis of segmental co-ordination was not performed. As suggested by Chaffin et al. (1999), the learning process associated with material handling devices is rather slow and requires many replications, or bouts of repetitive lifting. With this in mind, as well as the fact that previous investigations involving the PLAD have been limited to single, constrained lifts with a minimal number of repetitions, it is unknown whether any learning/changes in lift kinematics and co-ordination occur with use of PLAD during repetitive lifting. In other words, questions of whether or not use of the PLAD causes operators to adopt a different lifting style and technique remain unanswered. Furthermore, it unknown whether or not these adaptations can be viewed as a positive or negative attribute associated with use of the device. To address this, the current investigation was undertaken to document any kinematic adaptations (i.e., lifting posture, kinematics, and co-ordination) associated with use of the device and further understand its potential use as an ergonomic intervention.

3.3 Methods

3.3.1 Participants

Ten healthy male subjects volunteered to participate in the study. They were representative of a university population, and had a mean (SD) age of 22 (3.8) years, height of 1.83 (0.5) m and weight of 85.3 (8.7) kg. Prior to participation, subjects were required to sign informed consent (as approved by the Queen's University Research Ethics Board – Appendix A) as well as a Par-Q Health questionnaire (Appendix B). At the time of the study, all subjects were free of musculoskeletal injury with no reported history of low back pain.

3.3.2 Instrumentation

3.3.2.1 Motion Analysis

Subjects were instrumented with four Fastrak™ electromagnetic motion sensors, placed at the sternum, dorsally at T12/L1, on the left iliac crest, and laterally at the centre of mass of the right thigh (estimated as the midpoint of the segment, measured from the greater trochanter to the lateral condyle of the femur (from Winter, 1990)). The sensors were fixed at each landmark using TUF-Skin® spray and adhesive tape. For the sternum and T12/L1 location, tubular bandages (Mefix™) were placed over the sensors and wrapped securely around the torso in order to minimize motion artifact during testing. During data analysis, measurements recorded from the sternum sensor were found to be erroneous, indicative of a hardware malfunction; as such, these data were removed from the study.

Orientation data from the remaining sensors were used for a 2D kinematic analysis of the lifting task through calculation of absolute and relative joint motion about the knee, hip, and lumbar spine (Figure 3-1). Lumbar kinematics (angle and velocity) were calculated based on relative motion between the T12/L1 and pelvic sensors. Hip kinematics (angle and velocity) were based on the relative motion between the pelvis and thigh; these data were calculated using the sensors attached to the pelvis and the right thigh. Knee angle was defined as the included angle between the right thigh and shank; this angle was estimated using a method similar to that described in Albert (1999), that utilized the geometric law of cosines, subject anthropometric data (thigh length, shank length) and continuous measurement of the vertical height of the thigh sensor relative to the floor.

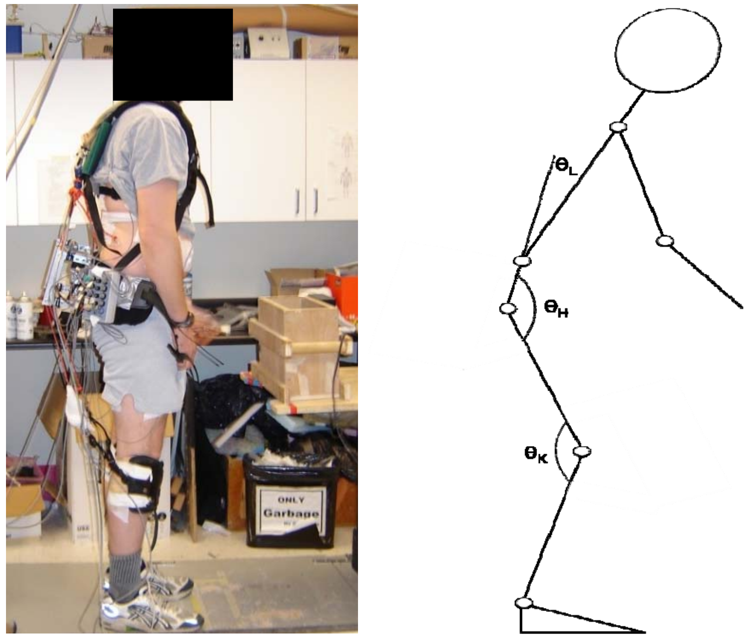

Figure 3-1 Left: Subject standing in erect posture prior to start of a repetitive lifting session used in the experiment. **Right:** Definition of joint angles used for the kinematic analysis, θ_L, θ_H, θ_K represent angles of the lumbar, hip, and knee, respectively.

3.3.3 Experimental Protocol

3.3.3.1 Orientation Session

Subjects were familiarized with the experimental setup and practiced lifting a light load paced to a metronome (6 lifts per minute). They then donned the PLAD system and were asked to continue practicing lifts while wearing the device. For this experiment, PLAD element stiffness was set to create forces (~180 N) similar to those reported in Abdoli et al. (2007). No preferred lifting style or technique was suggested to subjects: they were simply told to lift in a manner they considered "normal." Chaffin et al. (1999) suggested that motor learning associated with use of material handling devices is rather slow, requiring greater than 40 repetitions. As such, this training period was included to minimize learning effects during the repetitive lifting session. At the conclusion of the orientation session, subjects were restrained in an upright, isometric-strength-testing apparatus that featured an in-series load cell (Interface, Berkshire, UK) attached to a strap that was placed around each subjects torso, underneath the right and left axillas. Prior to testing, the load cell was calibrated using a set of known masses and a linear calibration function was determined (R^2 = 0.97) and then applied during trial tests in order to express voltage output in terms of force (N). Once in the apparatus, subjects were required to perform 3 maximum isometric trunk extensions. Maximum trunk strength (max force (MVC) obtained from load cell) quantified during these tests was subsequently used to scale the load to be lifted during subject testing (20% MVC).

3.3.3.2 Sessions 2 and 3

Subjects were required to participate in two sessions of repetitive lifting with and without use of the PLAD. Trial order was balanced and randomized across subjects. The task consisted of 9 blocks of lifting trials, with each trial block lasting 5 minutes, amounting to a total of 45 minutes of repetitive lifting. The lifting task required subjects to lift and lower a load in a box with handles from floor to knuckle height. The load was individually scaled to 20% of each subject's maximum (%MVC) as determined through an isometric strength test that had been conducted during the orientation session. Subjects were again instructed to lift freely using any desired technique. The task was paced using an audible tone (generated by computer) that subsequently guided the subjects to perform six lifts and six lowers per minute (30 lifts within each 5 minute trial block). Following the completion of each 5 minute trial, subjects were immediately placed in a testing apparatus, where they performed three isometric reference contractions. Reference contraction intensity was set to 50% MVC; performance feedback was provided to the subjects via a real-time display. Following the completion of the reference contractions, subjects were directed back to the lifting area for another lifting session. Prior to the start of each trial block, subjects were asked to assume a reference posture that was used to align the anatomical reference frame with the co-ordinate system of the motion analysis hardware (Fastrak™, Polemus, Colchester, VT). Kinematic data were collected continuously and stored throughout each bout of lifting. The EMG data collected during the reference contractions were used to quantify localized muscle fatigue in the low back musculature via spectral analysis and determination of the medial power frequency of the EMG signals; these data and their methods have been published elsewhere (Lotz et al., 2007). Briefly, the EMG results indicated no measurable indications of fatigue during the first three blocks of lifting. As such, data collected during the 15 minutes of repetitive lifting were used in the current study to directly assess the effects of the PLAD on lifting kinematics and co-ordination.

3.3.4 Data Processing

3.3.4.1 Extraction of Kinematic Waveforms

Data from the Fastrak™ sensors were digitally low-pass filtered using a 4th order Butterworth filter with a final cut-off frequency of 3 Hz. Joint angles and velocities were then determined from these data, yielding six kinematic waveforms that described motion of the knee, hip, and lumbar spine. Waveform data were further processed using custom software (LabVIEW 7.1, National Instruments, Austin, TX) designed to isolate and extract data from each of the 30 lifts that occurred within each 5-minute lifting trial. The custom software featured a graphical interface that enabled the researcher to identify and record the start and stop points for each lift within the trial data set (Figure 3-3). These points were identified based on analysis of lumbar joint angle and velocity. The start of a lift was defined as the point at which flexion of the trunk started and the angular velocity of the trunk deviated from zero; the end of a lift was defined as the point at which the velocity of the trunk returned to zero after placement of the load.

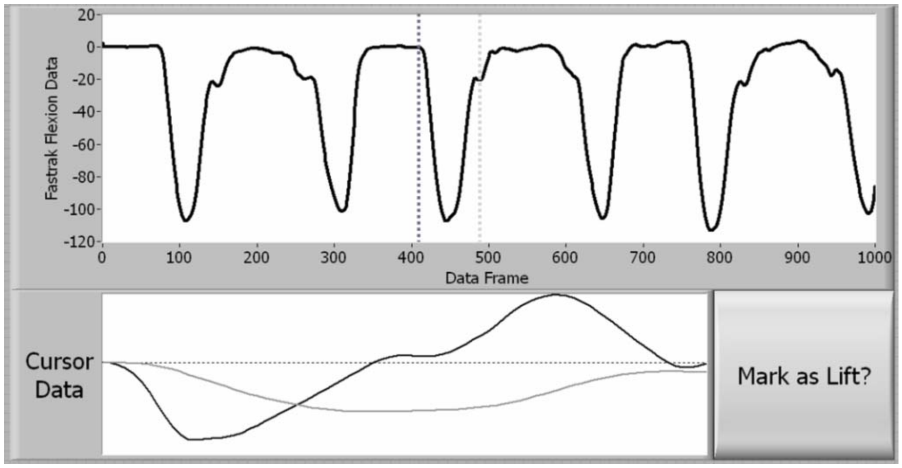

Figure 3-2 Graphical display of the custom software used to extract kinematic data pertaining to each individual lift.

3.3.4.2 Calculation of Relative Phase Angles

Co-ordination was evaluated through computation of relative phase angles as defined by Burgess-Limerick et al. (1993) and required several processing steps, which are depicted in Figure 3-3a,b,c and are follows: In Figure 3.3a, a single trial of lumbar hip and knee kinematic data is shown. From here, joint angle and velocity data pertaining to a lift were normalized to a range of +1 to -1, based on minimum and maximum values identified within each time series. The normalized data were then represented on a phase plane, with joint angle and joint velocity plotted on the X and Y axes respectively (Figure 3-3b). Each lifting cycle was characterized as a clockwise trace about the four quadrants of the phase plane. As such, the bottom quadrants represent joint flexion, describing the means in which subjects "approached" the load; the upper quadrants characterized joint extension and the actual lifting of the load. Phase angles (α) were determined via the inverse tangent of the normalized joint angle and velocity data projected on the phase plane. Relative phase angles between segments were subsequently quantified through subtraction of the distal joint phase angle from that of the proximal joint.

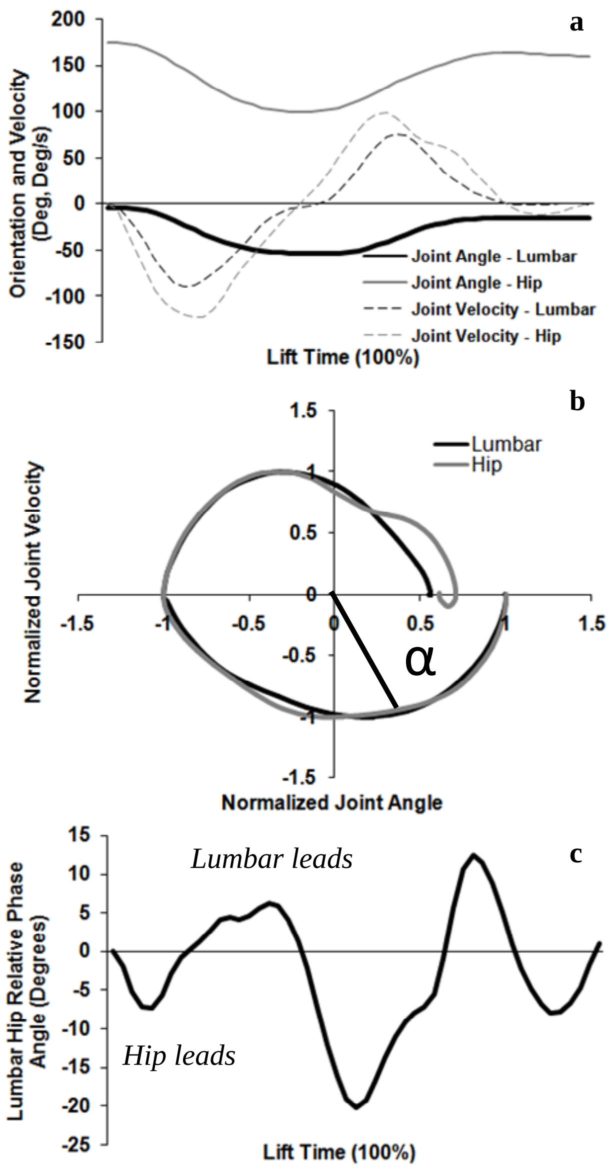

Figure 3-3a,b,c Processing steps required to quantify relative phase angle from a single lift. Joint angle and velocity data were extracted (**a**), normalized and projected onto a phase plane (**b**), and subtracted to calculate the relative phase angle between segments (**c**).

In this representation, a positive relative phase value indicated periods in the lifting cycle where the proximal joint led the distal in terms of relative motion (Figure 3-3c). Negative values indicated the opposite trend occurred. A phase angle of zero indicated complete synchrony between segments. Relative phase angles were obtained for every lift for the following segment pairs: lumbar-hip (L-H), lumbar-knee (L-K), and hip-knee (H-K).

3.3.4.3 Determination of Kinematic Variables and Statistical Analysis

For each kinematic waveform (i.e., three joint angle, three joint velocity, and three relative phase time series), a total of 90 curves were extracted for both the PLAD and No-PLAD conditions, yielding a grand total of 180 curves per subject. Using these data, several kinematic- and co-ordination-based parameters were quantified to statistically compare differences due to use of the PLAD. A listing of these variables is presented in Table 3-1. Values extracted from each individual curve were subsequently averaged to yield subject trial means within each of the lifting conditions. One-way repeated measures ANOVA tests ($\alpha = 0.05$) were used to assess the effects of the PLAD on each of the variables listed in Table 3-1. All statistical analyses were performed using SPSS 11.0 for Windows (SPSS Corporation, Chicago, IL). In addition, kinematic waveforms pertaining to each lift were time normalized to represent 100% of the lifting cycle and ensemble averaged across subjects to produce a representative curve for each kinematic variable. These data served to provide a graphical, qualitative account of the effects of the PLAD on lifting kinematics and co-ordination.

Table 3-1 Dependent variables used to describe lifting kinematics and co-ordination.

Kinematic Variables
Task
Lift Time (s)
Lumbar
Lumbar Flexion at Lift Onset (Deg)
Mean Flexion/Extension Velocity (Deg/s)
Peak Flexion/Extension Velocity (Deg/s)
Hip
Hip Flexion at Lift Onset (Deg)
Mean Flexion/Extension Velocity (Deg/s)
Peak Flexion/Extension Velocity (Deg/s)
Knee
Knee Flexion Lift Onset (Deg)
Mean Flexion/Extension Velocity (Deg/s)
Peak Flexion/Extension Velocity (Deg/s)
Co-ordination Variables
Lumbar-Hip (L-H)
Mean Flexion/Extension Relative Phase Angle (Deg)
Peak Flexion/Extension Relative Phase Angle (Deg)
Lumbar-Knee (L-K)
Mean Flexion/Extension Relative Phase Angle (Deg)
Peak Flexion/Extension Relative Phase Angle (Deg)
Hip-Knee (H-K)
Mean Flexion/Extension Relative Phase Angle (Deg)
Peak Flexion/Extension Relative Phase Angle (Deg)

3.4 Results

3.4.1.1 Lift Time and Posture at Lift Onset

Trial means (SD) and results of the repeated measures ANOVA are summarized in Table 3-2. Use of the PLAD did not significantly affect lifting time over the course of the repetitive lifting session ($\underline{F}$ (1, 9) = 0.187, p = 0.675). In terms of the lifting posture, use of the PLAD had a significant effect in terms of the amount of lumbar flexion ($\underline{F}$ (1, 9) = 5.94, p = 0.038) and hip flexion ($\underline{F}$ (1, 9) = 5.04, p = 0.05) utilized by subjects during pick up of the box. This equated to an ~9 degree decrease in peak trunk flexion and an ~5 degree increase in peak hip flexion angle associated with use of the PLAD over the course of repetitive lifting. The effect of the PLAD on knee flexion magnitude was not significant ($\underline{F}$ (1, 9) = 1.67, p = 0.228), although it should be noted that a large, yet variable difference existed in terms of the amount of knee flexion employed during the repetitive lifting task. Ensemble averages of the lifting time series are presented in Figure 3-4, detailing postural differences observed due to use of the PLAD.

Table 3-2 ANOVA results of lifting times and postures observed across experimental (No-PLAD/PLAD) conditions. Significant differences are highlighted in bold, italicized typeface.

Dependent Variable	Trial Means (SD) and ANOVA Test Results		
	No-PLAD	PLAD	*p*-value
Lift Time (s)	2.67 (0.33)	2.70 (0.32)	0.675
Lumbar Flexion (Deg)	***-48.51 (10.04)***	***-39.67 (10.71)***	***0.038***
Hip Flexion (Deg)	***86.81 (9.97)***	***82.2 (9.22)***	***0.05***
Knee Flexion (Deg)	92.5 (15.34)	84.9 (23.53)	0.228

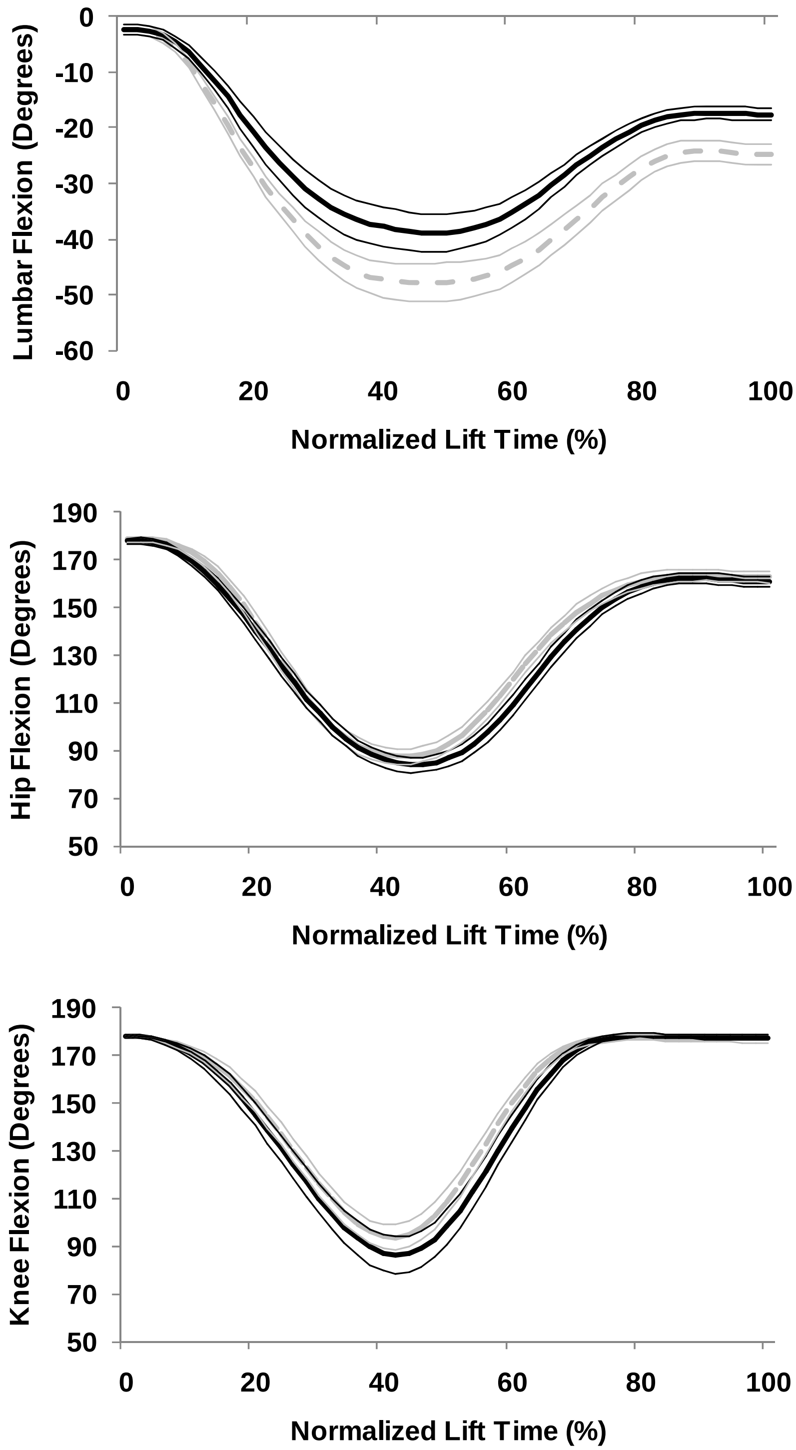

Figure 3-4 Ensemble averaged time series (across all subjects) describing lumbar, hip, and knee motion during lifting with (**black**) and without (**grey**) use of the PLAD. Thin lines represent +/- 1 SEM.

3.4.1.2 Flexion/Extension Velocity During Lifting

Trial means (SD) and results of the repeated measures ANOVA are summarized in Table 3-3. As subjects flexed forward to pick up the box, use of the PLAD significantly reduced both average and peak trunk flexion velocity (F (1, 9) = 8.195, p = 0.019, and F (1, 9) = 22.545, p = 0.001), respectively. A significant decrease in peak hip flexion velocity was also observed (F (1, 9) = 14.2, p = 0.004). Mean and peak knee flexion velocity did not significantly change with use of the PLAD.

During lifting and placement of the box on the shelf, minimal differences were found to exist between conditions with and without use of the PLAD. However, subjects were observed to lift with a greater mean hip extension velocity (F (1, 9) = 7.373, p = 0.024) by a magnitude of approximately 4 Deg/s. As shown in Table 3-3, a greater difference was observed between No-PLAD/PLAD conditions in terms of lumbar extension velocity; however, the large variability present in the PLAD condition prevented this difference from approaching statistical significance. As with knee posture and knee flexion velocity, no significant differences were found to occur due to use of the PLAD in regards to extension velocity about the knee. To summarize, use of the PLAD caused a decrease in the rate at which subjects flexed at the trunk (~6 Deg/s, ~20 Deg/s, peak and mean velocity, respectively) and hips (~20 Deg/s, peak velocity) as they approached the box prior to lifting. During lifting, use of the PLAD generally caused subjects to lift with significantly greater extension velocity about the hip. Ensemble averages of the lifting time series are presented in Figure 3-5, detailing changes in segmental flexion and extension velocity associated with use of the PLAD.

Table 3-3 ANOVA results of joint flexion/extension velocities observed across experimental (No-PLAD/PLAD) conditions. Significant differences are highlighted in bold, italicized typeface.

Dependent Variable	Trial Means (SD) and ANOVA Test Results		
	No-PLAD	PLAD	*p*-value
Flexion (Load Approach)			
Mean Lumbar Velocity (Deg/s)	***-32.66 (5.65)***	***-26.23 (10.26)***	***0.019***
Peak Lumbar Velocity (Deg/s)	***-85.11 (17.93)***	***-65.27 (27.55)***	***0.001***
Mean Hip Velocity (Deg/s)	-71.23 (21.29)	-68.26 (14.08)	0.353
Peak Hip Velocity (Deg/s)	***-171.6 (31.62)***	***-150.79 (27.56)***	***0.004***
Mean Knee Velocity (Deg/s)	-63.19 (10.48)	-67.91 (18.78)	0.368
Peak Knee Velocity (Deg/s)	-148.19 (25.72)	-146.07 (27.19)	0.731
Extension (Lift)			
Mean Lumbar Velocity (Deg/s)	20.71 (4.77)	20.02 (9.90)	0.780
Peak Lumbar Velocity (Deg/s)	44.5 (10.12)	39.93 (22.35)	0.241
Mean Hip Velocity (Deg/s)	***60.63 (9.83)***	***64.77 (8.25)***	***0.024***
Peak Hip Velocity (Deg/s)	137.29 (23.52)	135.54 (22.16)	0.748
Mean Knee Velocity (Deg/s)	70.92 (19.71)	77.17 (17.61)	0.331
Peak Knee Velocity (Deg/s)	176.36 (35.32)	181.63 (31.77)	0.617

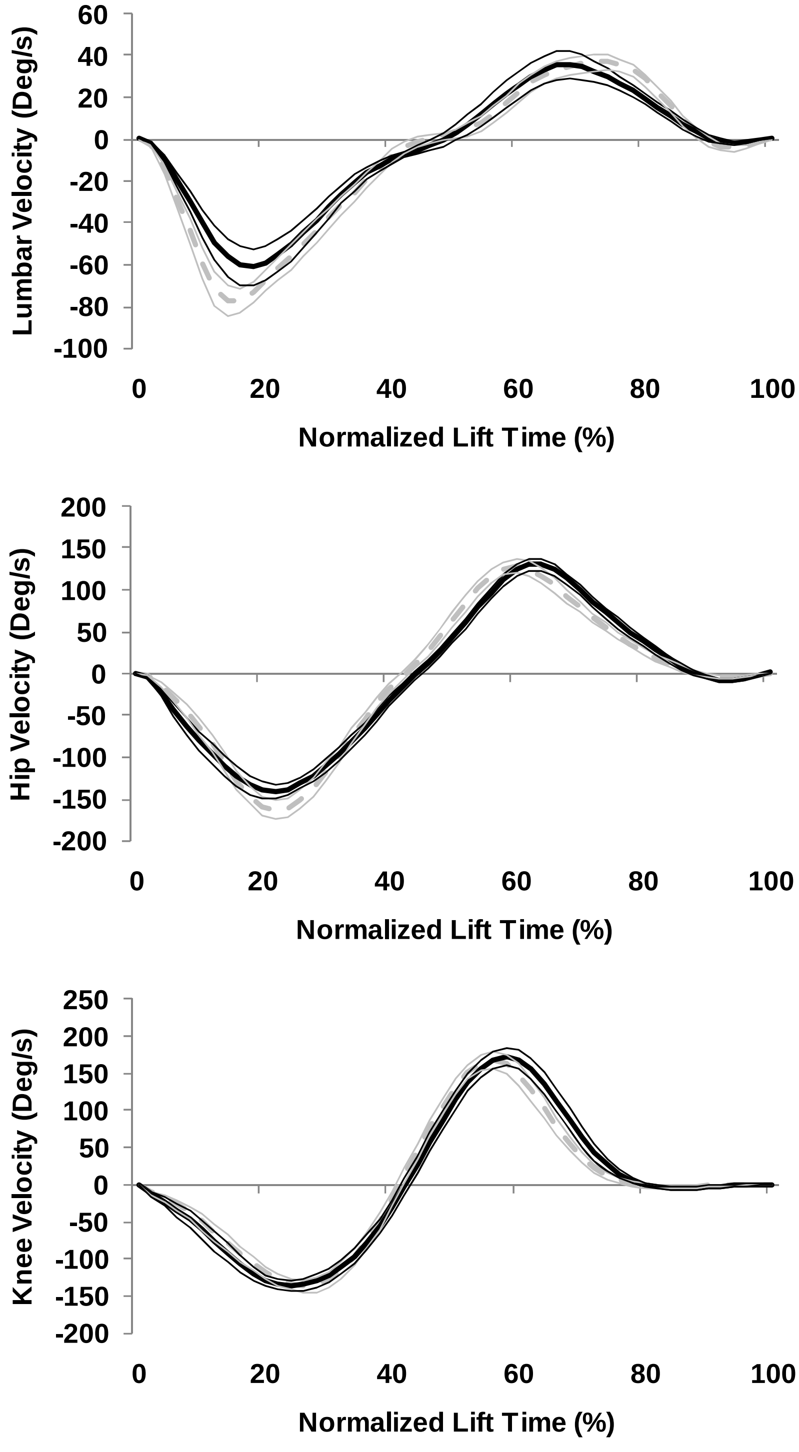

Figure 3-5 Ensemble averaged time series (across all subjects) describing lumbar, hip, and knee velocity during lifting with (**black**) and without (grey) use of the PLAD. Thin lines represent +/- 1 SEM.

3.4.1.3 Relative Phase Angles

Trial means (SD) and results of the repeated measures ANOVA are summarized in Table 3-4. During approach and subsequent lifting of the load, use of the PLAD significantly altered co-ordination between the lumbar–hip (L-H) segments, as well as the lumbar-knee (L-K), both in terms of mean and peak relative phase angle magnitude. As subjects approached the load while wearing the PLAD, peak and mean relative L-H phase angles decreased by ~10 and ~8 degrees, respectively ($\underline{F}$ (1, 9) = 10.97, p = 0.009; $\underline{F}$ (1, 9) = 11.622, p = 0.008). Across the lumbar and knee segments (L-K), similar results were observed, as use of the PLAD significantly reduced peak and relative phase angles by magnitudes of ~12 and ~11 degrees ($\underline{F}$ (1, 9) = 7.29, p = 0.024; $\underline{F}$ (1, 9) = 8.747, p = 0.016). No significant differences were observed with respect to co-ordination between the hip and knee (H-K), ($\underline{F}$ (1, 9) = 0.239, p = 0.637 for peak values, and $\underline{F}$ (1, 9) = 0.209, p = 0.659, for mean values).

As subjects lifted using the PLAD, peak and relative L-H phase angles decreased by ~16 and ~12, respectively ($\underline{F}$ (1, 9) = 6.438, p = 0.032; $\underline{F}$ (1, 9) = 11.555, p = 0.008). In terms of relative motion between the lumbar and knee (L-K), lifting with the PLAD caused significant decreases in relative phase angle magnitude, with peak and mean differences estimated to be ~14 and ~11 degrees, respectively ($\underline{F}$ (1, 9) = 12.75, p = 0.006; $\underline{F}$ (1, 9) = 12.92, p = 0.006). No significant differences were observed with respect to co-ordination between the hip and knee (H-K), ($\underline{F}$ (1, 9) = 0.243, p = 0.634; $\underline{F}$ (1, 9) = 1.16, p = 0.308, peak and mean values, respectively). Ensemble averages of the lifting time series are presented in Figure 3-6. As shown in the Figure, a proximo-distal sequence (positive phase angle) of rotations was evident as subjects flexed forward to pick up the box. During lifting (extension), a disto-proximal sequence of rotations was observed (negative phase angle).

Table 3-4 ANOVA results of relative phase angles observed across experimental (No-PLAD/PLAD) conditions. Significant differences are highlighted in bold, italicized typeface.

Dependent Variable	Trial Means (SD) and ANOVA Test Results		
	No-PLAD	PLAD	*p*-value
Flexion (Load Approach)			
Mean L-H Relative Phase Angle (Deg)	***15.04 (7.87)***	***7.64 (10.61)***	***0.008***
Peak L-H Relative Phase Angle (Deg)	***30.15 (14.19)***	***20.11 (14.12)***	***0.009***
Mean L-K Relative Phase Angle (Deg)	***20.24 (8.64)***	***11.65 (13.24)***	***0.016***
Peak L-K Relative Phase Angle (Deg)	***37.57 (14.13)***	***25.39 (16.13)***	***0.024***
Mean H-K Relative Phase Angle (Deg)	5.19 (8.05)	4.01 (6.88)	0.659
Peak H-K Relative Phase Angle (Deg)	15.08 (8.26)	13.39 (7.58)	0.637
Extension (Lift)			
Mean L-H Relative Phase Angle (Deg)	***-41.61 (7.81)***	***-29.61 (12.72)***	***0.008***
Peak L-H Relative Phase Angle (Deg)	***-78.23 (16.01)***	***-62.89 (24.39)***	***0.032***
Mean L-K Relative Phase Angle (Deg)	***-62.63 (9.51)***	***-51.92 (9.27)***	***0.006***
Peak L-K Relative Phase Angle (Deg)	***-110.62 (13.71)***	***-96.04 (13.51)***	***0.006***
Mean H-K Relative Phase Angle (Deg)	-21.02 (4.25)	-22.31 (6.17)	0.308
Peak H-K Relative Phase Angle (Deg)	-49.65 (10.11)	-51.02 (12.39)	0.634

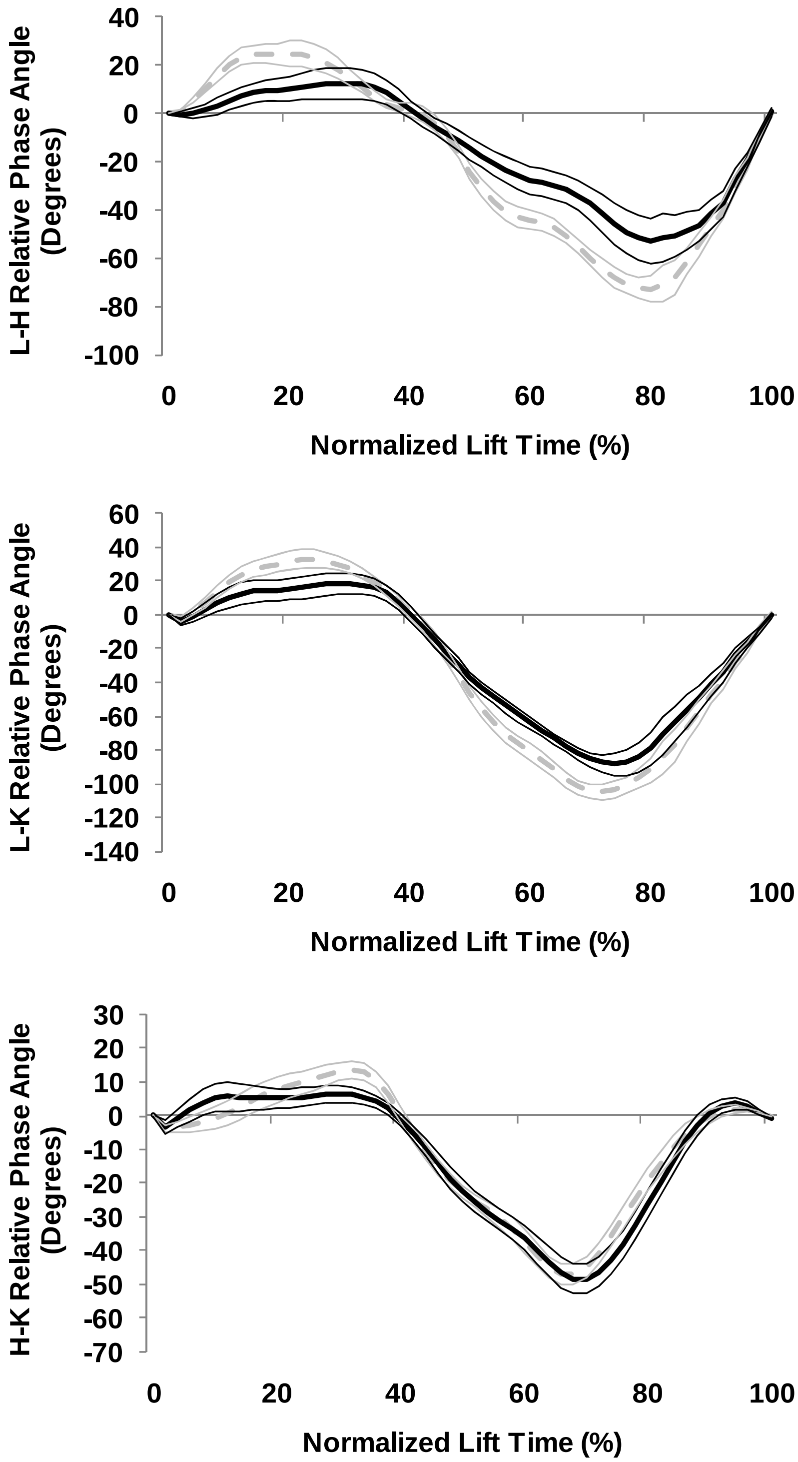

Figure 3-6 Ensemble averaged time series (across all subjects) describing lumbar, hip and knee relative phase relationships (co-ordination) during lifting with (**black**) and without (**grey**) use of the PLAD. Thin lines represent +/- 1 SEM.

3.5 Discussion

The purpose of the current study was to further our understanding of how an assistive device, like the PLAD, affects segmental kinematics and co-ordination during lifting. To do so, subjects were required to perform a repetitive sagittal lifting task with and without use of the assistive device. Subjects were encouraged to lift freely, using any style they preferred. Kinematic data obtained from Fastrak™ motion sensors were used to quantify joint angles, velocities, and relative phase angle relationships between segments. Based on the results obtained, it is apparent that use of the PLAD had a significant effect on lifting posture, joint velocity, and co-ordination across joints.

At box pick-up, use of the PLAD device caused subjects to assume a lifting posture that utilized significantly less lumbar flexion and significantly more hip flexion when compared to the No-PLAD condition (Figure 3-4). On average, these changes were quantified to be an ~9 degree decrease in peak lumbar flexion, and an ~5 degree increase in flexion about the hip. This general trend has been observed repeatedly in research regarding elastic lifting belts and corsets, and comparable results in terms of changes in lumbar and hip flexion have been reported in the literature (Granata et al, 1997; McGorry & Hsiang, 1999; Sparto et al., 1998).

Excessive trunk flexion has been identified as a risk factor for low back pain (Marras et al., 1995; Norman et al, 1998), and it naturally causes increased joint loading; this is due to the moment caused by the upper body mass, as well as those posed by external forces or loads. From an aetiological perspective, lumbar spine flexion has been shown to modify the strength-bearing capacity of lumbar vertebrae and intervertebral discs (Gunning, Callaghan and McGill, 2001; Callaghan and McGill, 2001) and place additional strain on the ligaments and posterior elements of the vertebral bodies (i.e., facet joints). Furthermore, towards full lumbar flexion, the erector spinae lose their oblique line of action, diminishing, not only their moment-generating capacity but also their ability to oppose anterior shear forces (Potvin, Norman, and McGill, 1991). As such, practitioners have encouraged the maintenance of a neutral spine posture during lifting and suggest forward flexion of the trunk through pelvic tilt and use of the “hip hinge” (McGill, 2002). The results obtained in this experiment suggest use of the PLAD encourages this type of lifting, and as such, may influence operators to lift with more of an “optimal” spine posture.

During the initial phase of lifting, use of the PLAD system caused subjects to approach the box with significantly reduced flexion velocity (Figure 3-5). This was quantified about both the lumbar and hip joints. This change was most likely due to the passive stiffness of the PLAD; as subjects flexed forward, the PLAD elements were stretched, creating a resistive moment that opposed forward flexion, leading to a decrease in the flexion velocity of the trunk and hip. Despite this decrease, no differences were observed in the time required to complete each lifting cycle. This was, in part, explained by the significantly larger mean hip extension velocity observed during lifting with the PLAD. While the resistance of the PLAD did not affect performance (i.e., task lifting time) or trunk extension velocity in this experiment, it is unknown whether or not this response would be evident with lifting tasks at higher frequencies or with similarly paced tasks featuring use of a stiffer PLAD element. Fast-paced occupational tasks requiring large trunk velocities have been identified as “high risk” in terms of injury generating potential (Marras et al., 1995). This was suggested to be a consequence of higher spinal loads, increased co-contraction, and segment/load inertia. Recently, Granata et al. (2006) suggested that neuromuscular control becomes less stable with increasing trunk velocity, leading to an increased likelihood of injury due to motor error or instability. With this in mind, the decrease in flexion velocity associated with use of the PLAD can be viewed as beneficial, so long as it does not impede task performance, causing significantly greater trunk extension velocities during lifting in an attempt to “make up” time spent flexing forward against the resistance of the PLAD elements.

In terms of co-ordination (synchrony) between segments, the PLAD was found to significantly affect the phase lag exhibited between the lumbar spine and the hip, and between the lumbar spine and the knee (Figure 3-6). This trend was evident both as subjects approached and lifted the box. As subjects flexed forward towards the box, a proximo-distal sequence of rotations was observed: in other words, the trunk led the hip and the knee. During lifting of the box, the opposite trend occurred. This behavior is consistent with literature describing co-ordination in lifting (Sparto et al., 1998; van Dieen et al., 1996; Lindbeck & Kjellberg, 2001; Burgess-Limerick et al., 1993; McGorry & Hsiang, 1999). While a significant change in relative phase (co-ordination/synchrony) was observed, this does not imply that use of the PLAD facilitates users to lift with "better" or "increased" co-ordination and technique. The decrease in relative phase angle due to use of the PLAD suggests more synchronous movement between segments. Because the PLAD elements attach at the shoulder and below the knee, they cross each of these joints and are "tri-articular" in a sense. As tension develops in the PLAD elements, it is possible that this tension causes the increased synchrony that was observed during flexing motions across these joints. However, this does not explain the observed changes during lifting of the box. Relative phase angles have been suggested to be measures reflective of a motor control strategy employed to optimize energy transfer throughout a kinetic chain. As such, changes in these angles are thought to correspond with a control strategy that keeps bi-articular muscles at their maximum contractile efficiency, by enabling a given muscle to remain at an optimal length and minimizing contraction velocity (Burgess-Limerick et al., 1995).

Accordingly, the relative phase angle between the knee and lumbar spine has been shown to increase proportionately with heavier loads (Scholz, 1993; Burgess-Limerick et al., 1995). It is suggested that this increased lag was meant to keep the trunk musculature at its optimum strength-generating capacity when the effects of load acceleration were greatest, thus reducing the demands placed on the trunk muscles (Burgess-Limerick et al., 1995). When subjects lifted the box while wearing the PLAD, the relative phase angles for both the lumbar-knee (L-K) and lumbar-hip (L-H) segments significantly decreased. This response could be due to the observed changes in lifting posture at lift onset and/or the mechanical assistance provided by the PLAD. Previous research has suggested that at the onset of lifting an object up from the ground, the PLAD can generate 20-25 Nm of extensor moment about L4/L5 (Abdoli et al., 2007). Despite the fact that the current study utilized a PLAD of newer design, its force-generating capacity was kept similar to what has been reported in previous work. With this in mind, it could be argued that the decreases in relative phase angle observed here reflect a reduction in the "effective" biomechanical demand actually placed on the trunk muscles during lifting. In other words, compared to the No-PLAD condition, the assistive moment provided by the PLAD caused users to exhibit a co-ordination strategy that would comparatively be associated with lifting of a lighter load. While these results support the notion that the PLAD system does provide assistive support to the trunk musculature and "take up part of the load", they could pose serious consequences. For example, it is unknown whether the effects of the device will have a lasting effect on co-ordination, particularly during any type of lifting, or material handling that immediately follows an extensive period of use of the device. As such, further research is needed to investigate the long-term effects of PLAD and to determine if there are any carry-over effects that might exist when the device is no longer worn.

In summary, the goal of this study was to quantify changes in lifting kinematics and co-ordination associated with use of the PLAD. During a repetitive lifting task, use of the PLAD caused subjects to exhibit lifting postures that featured less flexion at the lumbar spine and more flexion at the hips. Furthermore, use of the PLAD caused significant changes in segmental co-ordination during lifting, particularly related to co-ordination of movement between the lumbar spine and the hips and between the lumbar spine and the knees. These findings are in disagreement with previous research that observed no significant changes in lift kinematics due to use of the PLAD (Abdoli, et al., 2006). This is most likely due to differences in the methodologies employed. In the current study, participants were provided with an extended preliminary training session to get accustomed to lifting with the PLAD prior to data collection. A preliminary session was not provided in Abdoli et al. (2006), and subjects were not required to perform repetitive lifts. As evidenced by Chaffin et al. (1999), the motor learning process associated with use of material handling devices is rather slow, requiring greater than 40 repetitions. Following the training session, the data collection session required subjects to perform an additional 90 lifts with the PLAD using their own preferred lifting style. In Abdoli et al. (2006), participants were required to perform a series of 1-repetition lifts that were presented in sequential order, randomized by lifting style, and load magnitude. Because of this, it is likely that participants in Abdoli et al. (2006) did not have a sufficient opportunity to learn how to adapt their lift kinematics in response to the assistive support provided by the device, and as such, lifting kinematics remained invariant when the PLAD was used. As the current study provided sufficient practice and repetitions, it is clear that with sufficient practice, use of the PLAD device has an appreciable effect on lifting posture, kinematics and co-ordination.

While the findings do provide further understanding regarding adaptations due to use of the PLAD, they are not without limitation. The current study utilized a symmetric, sagittal lifting task preformed at a set frequency using a load scaled to 20% MVC. It is unknown whether these results are generalizable toward lifting tasks involving asymmetric postures, different lift frequencies, and/or load mass; as such, further investigations are warranted. In addition, the dependent variables used within the study were general parameters of curves, and in many instances, only described a discrete component or period of the lifting task. While an attempt was made to account for this by way of assessing both the approach, and lifting components of the task, again these data serve as relatively crude, discrete parameters that can be used to describe a curve, or waveform. To build on these findings, more advanced analysis techniques, such as principal component, or functional analyses of variance (fANOVA) should be considered, as recent evidence has illustrated these methodologies to provide further insight simply beyond single curve parameters, such as mean values, peaks, etc (Godwin, et al., in press). Furthermore, future investigations should also be aimed at assessing whether similar effects are evident with females, who have been indentified to naturally lift with different technique than males (Lindbeck et al., 2001). Finally, it is critical that future research investigates the long-term impacts of using the device, in an attempt to determine the potential for of this ergonomic aid to be used as a safe and effective engineering control for manual materials handling tasks.

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Chapter 4
The Effect of a Personal Lift Assist Device (PLAD) on Active Trunk Stiffness: Implications for Spine Stability

4.1 Abstract

Stability of the trunk is modulated by biomechanical loads, postures, and neuromuscular control strategies. In regards to biomechanical or kinetic factors, the active stiffness contribution toward spine stability increases both as a function of joint compressive load and muscle force. In previous research investigating the effectiveness of the PLAD, significant decreases in both trunk muscle activity and compressive spine loads have been documented. Based on this knowledge, it is possible that using the PLAD device will have an impact on trunk stiffness as well. Therefore, the purpose of the current study was to quantify the passive rotational stiffness of the PLAD system and assess the absolute and relative effects of the PLAD on active trunk stiffness. Nine male subjects participated in the study. Subjects were required to assume a series of static, symmetrical flexed postures, varied by trunk flexion magnitude. Within each posture, a No-PLAD condition was tested, as were the effects of five different levels of PLAD rotational stiffness. Normalized electromyographic (EMG) data were quantified across seven muscles on the right side of the body. Bilateral symmetry was assumed. EMG, trunk posture, and PLAD stiffness data were input into an EMG-assisted stability model that estimated active trunk stiffness, based on the input data and anatomical and geometric constraints. Around neutral trunk posture, up to 15 degrees of trunk flexion, the results suggest the PLAD increases the overall stiffness of the trunk. However, use of the PLAD significantly compromised the active stiffness of the trunk as flexion increased (21-46% decrease). This effect was consistent across all PLAD conditions. Based on results, use of the PLAD, with its current design, may place workers at an increased risk of injury in events where high sudden loads or unexpected perturbations are

applied to the trunk. Future research is recommended, as are design changes regarding the function of the PLAD and its components.

4.2 Introduction

Operators working in material handling and packaging environments are constantly required to move or displace objects that vary by shape and size. Interaction with these objects typically involves lifting and lowering exertions, as well as pushing or pulling efforts, and may include use of an assistive device. Whether due to mislabeling, a lack of labeling, or instances where an operator incorrectly judges the mass of the object being displaced, manual material handling tasks are inherently prone to providing instances that can cause sudden loads, or perturbations to be applied to joints of the body, including the low back. The body's ability to prepare or respond to events like these is governed by the neuromuscular control system's ability to actively stabilize the system (via muscle activation and reflexive control) and by the passive stiffness provided by the joint(s) to which the sudden load or perturbation is applied. It is theorized that in instances where the sudden change in "task demand" is greater than expected, excessive displacement can occur between adjacent anatomical structures (e.g., buckling between vertebrae), which places high strain on supporting structures, ultimately causing them to become injured (McGill, 2002). Using non-linear systems theory, Granata et al. (2006) recently demonstrated that neuromuscular control of the trunk becomes less stable with increased trunk velocity. As trunk velocity has been shown to be a predictor for LBP risk (Marras et al., 1995), these findings correlate with the notion that injury caused by instability can occur in the workplace.

Stability of the trunk is modulated by biomechanical loads, postures, and neuromuscular (reflexive) control strategies (Granata et al., 2006). In regards to biomechanical or kinetic factors, the active stiffness contribution toward spine stability increases both as a function of joint compressive load and muscle force (Stokes and Gardner-Morse, 2003). Research investigating the effects of lifting belts and lumbosacroal orthoses suggests that the additional passive stiffness supplied by these devices add stability to the trunk (McGill, 1990, Cholewicki et al., 1999, Ivancic et al., 2002). Furthermore, this research suggests that the additional support supplied by these devices causes mild reduction in activation levels of the trunk extensor musculature, ultimately leading to a moderate decrease in the compressive loads incurred in the lumbar spine. For the last few years, research at our laboratory has focused on developing an ergonomic lifting aid that exhibits a similar response. The Personal Lift Assist Device (PLAD) was designed to provide assistive support to the trunk extensor musculature during lifting tasks. To date, we have demonstrated its effectiveness at reducing erector spinae muscle demand in single and repetitive lifting tasks (Abdoli et al., 2006, 2007; Lotz et al., 2007) and have theorized its ability to reduce compression and shear forces acting on the lumbar spine (Abdoli et al., 2007). As active trunk stiffness increases as a function of both joint load and muscle force, it is reasonable to assume that use of the PLAD will have an appreciable effect on active trunk stiffness, and as a result, trunk stability; however the extent and magnitude to which this relationship exists is unknown. As such, the purpose of this study was to assess the effect of the PLAD on trunk stiffness and the implications of its potential use as an ergonomic intervention.

4.3 Methods

4.3.1 Participants

Nine healthy male subjects volunteered to participate in the study. Subjects were from a university population with a mean (SD) age of 21.7 (0.9) years, height of 1.81 (0.05) m and a mass of 81.3 (8.1) kg. When asked, all subjects indicated that they were free of musculoskeletal impairment and had not experienced any previous back, hip, or knee problems in their recent medical history. Subjects were required to sign informed consent (as approved by the Queen's University Research Ethics Board) and complete a PAR-Q Health questionnaire (Appendix B).

4.3.2 Instrumentation

4.3.2.1 EMG Data

Bipolar Ag/AgCl EMG electrodes (Medi-Trace) were placed on the skin, with an inter-electrode distance of 2.5 cm, at seven different locations on the right side of the body in order to record signals from the following muscles: rectus abdominus (RA), external oblique (EO), internal oblique (IO), latissimus dorsi (LD), thoracic erector spinae (TES), lumbar erector spinae (LES), and multifidus (MULT). Prior to the attachment of the electrodes, the skin over the muscle prepared was cleaned and abraded with alcohol. Electrodes were placed according to the procedures described by Cholewicki and McGill (1996). For each experimental condition, bilateral symmetry was assumed. Therefore, right-side EMG signals were used to represent both the ipsi- and contralateral musculature. The assumption of equivalent right/left muscle activations were deemed valid, as this response has been identified in previous research involving sagittal lifting (Potvin et al., 1996), including earlier investigations of the PLAD (Abdoli et al., 2006).

Muscle signals were scaled with respect to a series of maximum voluntary contractions in order to normalize activation levels across subjects (%MVC). MVC trials required subjects to maximally contract specific muscles for 3-5 seconds while adopting a series of postures designed to isolate each of the seven muscles. Back extensor MVCs (TES, LES, MULT) were collected while the subject lay prone on a table, secured at the pelvis and legs. Subjects then performed several isometric extension exertions against resistance provided by the experimenter. Each test was repeated three times, and adequate rest was provided between trials in order to avoid the effects of fatigue. Flexor MVCs required subjects to lie supine with their knees bent and feet flat on the testing table. For the RA, subjects performed an isometric abdominal crunch against the resistance of the experimenter. For the remaining two flexor muscles (EO, IO), subjects again performed an abdominal crunch exercise; however, they were asked to couple the crunch with a slight twisting motion (left and right) in accordance with the oblique action lines of the musculature. For the final test, the subject sat on the bench and performed a resisted "pull-down" exercise targeted to activate the LD. The subject sat with his upper arm abducted and elbow flexed to 90 degrees and pushed down while the experimenter resisted the downward motion.

Throughout the experiment, EMG signals were amplified (gain = 1K-5K; input impedance = 10GΩ; CMRR = 115 dB at 60Hz, Bortec Octopus AMT-8, Calgary, Canada) and digitally recorded using a 12-bit A/D card (National Instruments, Austin, TX, USA) set at a sampling frequency of 1024 Hz. The raw EMG data were then digitally band-pass filtered (30-450 Hz) and stored for post processing. While an upper frequency cut-off of 450 Hz is common within the literature, a recent research article has recommended a lower cut-off of 30 Hz in order to remove heart rate artifact that has been shown to contaminate trunk EMG signals (Drake and Callaghan, 2006).

4.3.2.2 PLAD Device and PLAD Elements

Five different elements of varying stiffness were created for use with the PLAD during this study. The elements were made out of elastic material similar to that used in previous design iterations (JumpStretch™ as cited in Frost et al. (2006)). The length of the elements was determined based on consultation with a professional who suggested that elastic materials (such as those used within the PLAD) respond most reliably through a strain of 50-100% (Riazi, personal communication). Based on these suggestions and the expected elongation of the elements (Abdoli, 2007), each element was made to be 20 cm in length. Additionally, the five elements were also designed to generate forces that ranged between 50-150% of what has been observed in previous investigations of the device (Abdoli, et al., 2007, Lotz et al., 2007).

4.3.2.3 Determination of PLAD Element Stiffness

The stiffness of each set of PLAD elements (K1-K5) was determined from stress/strain tests performed using a custom-built apparatus (Figure 4-1). The apparatus consisted of a load cell (Interface, Berkshire, UK) and a displacement transducer (Unimeasure, Corvalis, OR). Using a plumb line, the displacement transducer was vertically aligned with the tensile axis of the load cell. Prior to testing, voltage signals recorded from the load cell and displacement transducer were linearly calibrated using a series of known masses and displacements and a line of best fit (R^2 = 0.97).

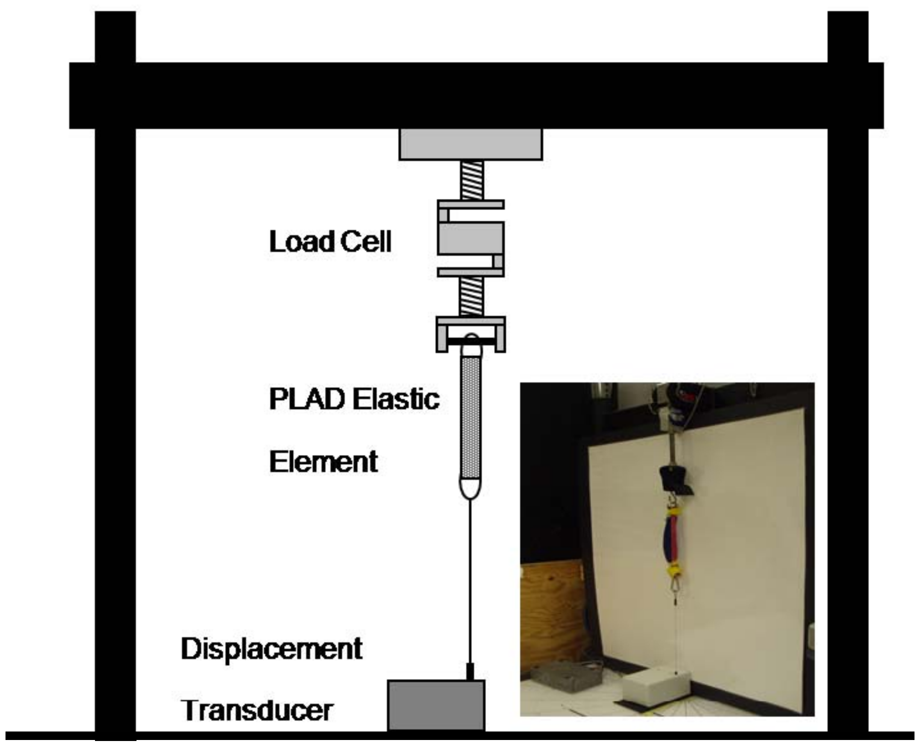

Figure 4-1 Custom apparatus (photo inset) used to quantify PLAD element stiffness.

To estimate the stiffness of the PLAD elements, each element was placed in the test apparatus and slowly stretched (manually) to 100% strain (~20cm). Each collection lasted 20 seconds and was repeated three times for each PLAD element, yielding a total of 15 trials. During these trials, load cell and displacement transducer data were captured at 1024 Hz using a 12-bit A/D card (National Instruments, Austin, TX, USA) and stored for post processing. Force and displacement data for each trial were subsequently smoothed using a 4^{th} order dual pass Butterworth filter with a final cut-off frequency of 3 Hz. The displacement-force curves obtained for each PLAD element were fit using 2^{nd} order polynomials. Differentiation of these polynomials produced equations that facilitated estimation of the "instantaneous" compliance of each element as well as the inverse of compliance: stiffness. These equations were developed due to their ease of use with data from the experimental trials, as PLAD element force was monitored, and not the change in length of the elements.

4.3.2.4 PLAD Load Cell

For each trial, the force generated by the PLAD elements was monitored through use of a load cell placed in series with the PLAD element. The load cell was custom made, and consisted of a strain gauge attached to an aluminum plate. Prior to testing, the load cell was calibrated using a set of known masses and use of a linear regression function, as the response of the load cell was found to be consistently linear (R^2 = 0.90 to 0.98 over repeated tests). During the experiment, voltages from the load cell were appropriately gained using a DC Amplifier (RDP Group, Modular 600), digitally sampled (along with the EMG data) at 1024 Hz using a 12-bit A/D card (National Instruments, Austin, TX), and stored for post processing. Force signals were subsequently doubled in amplitude to reflect the combined forces in both the right- and left-side elastic elements, and thus, the total stiffness that would actually be exhibited by the PLAD.

4.3.2.5 Motion Analysis Hardware

Trunk flexion was quantified using a Fastrak™ electromagnetic motion analysis system (Polhemus, Colchester, VT). Prior to the data collection, Fastrak™ motion sensors were placed on the sternum and over the left iliac crest and secured using TUF-Skin® spray and adhesive tape. Tubular bandages (Mefix™) were tied around the chest to assist in attaching the Fastrak™ sensor. Fastrak™ data were normalized with respect to an anatomical reference posture using the "boresight" function within the Fastrak™ hardware. This was necessary in order to estimate trunk flexion angle. The relative angular displacement between the sensors was quantified and subsequently used to represent flexion across the lumbar and thoracic regions of the spine. These data were collected in real time (30Hz) and presented to the subjects and experimenters via a graphical display created using custom software (LabVIEW 7.1, National Instruments, Austin, TX).

4.3.3 Experimental Protocol

Subjects performed 30 trials varied by flexion angle and PLAD stiffness. Trial conditions required subjects to bend forward to a predetermined amount of trunk flexion (15, 30, 45, 60, or 75 degrees). Biofeedback (trunk flexion) was presented to a computer monitor in order to help subjects reach the desired amount of trunk flexion. To ensure consistency in terms of the posture of the trunk and upper limbs, subjects were instructed to let their arms hang freely from their shoulders while holding on to a small (0.20m*0.15m*0.1m), empty cardboard container. Within each postural condition, subjects were required to wear the PLAD system, configured to one of six different levels of assistive stiffness. The six levels of PLAD stiffness tested included a No-PLAD condition as well as five PLAD (K1-K5) conditions that were varied by element stiffness. During the No-PLAD condition, it should be noted that subjects continued to wear the PLAD device; however, the shoulder and hip components were disconnected, preventing the device from having any mechanical effect. Trial order was randomized in blocks of element stiffness. Within each block, the order of flexion angle was also randomized. Data collection began once the subject reached the required flexion angle and remained motionless (as determined from the biofeedback). Once the desired amount of trunk flexion had been reached, the subject was instructed to hold that posture and remain motionless for five seconds. A scaled digital photograph was taken immediately following the collection period while the subject remained in the same posture. To reduce the likelihood of fatigue onset, short rest periods between each trial were given, along with longer breaks that naturally occurred during the changing of the PLAD elements. Between trials, the experimenter visually inspected the PLAD to ensure that it sat properly on the shoulders, hips, and knees; adjustments were made when necessary. Trunk posture, biomechanical, and EMG data collected during these trials were input into an EMG-assisted biomechanical spine model (Brown and Potvin, 2005) to predict the muscle forces required to maintain equilibrium. Muscle force and geometric data were then extracted from the model and input into a series of equations used to estimate trunk stiffness and the effect of the PLAD system.

4.3.4 Data Processing and Analysis

4.3.4.1 EMG Processing

Within each postural trial, muscle activity was quantified by calculating the root mean square (RMS) amplitude of each muscle signal. Maximum EMG amplitudes from the MVC trials were determined by applying a 150 ms sliding window to the EMG data and identifying the window containing the largest RMS amplitude. In the static flexion trials, the mean RMS amplitude of the trial was used to represent the activation state of each muscle. Muscle signals were then normalized with respect to the maximum RMS amplitudes observed during the MVC contraction trials and expressed in %MVC.

4.3.4.2 Estimation of External Moments and Centre of Mass

It was necessary to calculate the external moments acting at L4/L5 for each trial. In order to calculate the external demand, a scaled digital photograph was taken for each posture. Using biomechanical software (Q-Back, Queen's University), the subjects' postures were digitized (hand, wrist, elbow, shoulder, ear canal, C7/T1 and L4/L5) and input into the model (accompanied by subjects' anthropometric data) to calculate the net moment acting about L4/L5, as well as the height of the upper body centre of mass with respect to L4/L5. To minimize error, each posture was digitized three times and the outputs were subsequently averaged.

4.3.4.3 Estimation of PLAD Moment

PLAD moments were estimated by incorporating data obtained from the digital photograph and the in-series load cell. From each digital photograph, 2D vectors describing PLAD force ($\vec{F}_{PL}$) and moment-generating potential ($\vec{r}$) were determined. These vectors were calculated by digitizing the location of the L4/L5 joint as well as the PLAD element line of action running from the shoulder to the dorsal projection of the PLAD waist belt. The flexion/extension moment generated by the PLAD device was determined from the cross product of these vectors. The effective flexion extension moment arm of the PLAD (r_z) was also quantified. To minimize the influence of digitization error on these calculations, this process was repeated three times and the data were averaged.

4.3.4.4 EMG-Assisted Biomechanical Model

For each condition, estimates of muscle force were made through use of an EMG-assisted biomechanical model (Brown & Potvin, 2005). In short, the anatomical model consisted of a pelvis, lumbar vertebrae, a rib cage, 58 (29 right, 29 left) muscle fascicles, and their associated lines of action (from data of Cholewicki & McGill, 1996). Muscles represented in the model included the latissimus dorsi (LD), thoracic erector spinae (TES), lumbar erector spinae (LES), multifidus (MULT), rectus abdominus (RA), external oblique (EO), internal oblique (IO), as well as the Quadratus Lumborum (QL) and Psoas (PS) musculature. EMG data obtained during the study were used to bilaterally drive each respective muscle group. Activation for the QL and PS muscles were determined from the LES and IO signals, respectively, as suggested in Cholewicki and McGill (1996). For each trial, external L4/L5 moments were calculated using the scaled digital photograph and biomechanical modeling software (Q-Back). The passive moment contributions of the L4/L5 disc (McGill, 1994) and the PLAD system were then calculated and accounted for, leaving an estimate of the residual net moment opposed by the musculature.

Moment balance was then achieved for each posture by quantifying the muscle forces required to maintain static equilibrium, accounting for muscle activation, moment arm, muscle length, and cross sectional area. Calculations of individual muscle force were made using:

$$F_m = Gain[EMG_m * PCSA_m * \delta * L_m + F_p] \qquad [1]$$

Where F_m = muscle force (N), EMG_m = normalized EMG level (%MVC), $PCSA_m$ = muscle physiological cross-sectional area (cm^2), δ = maximum muscle stress, set at 35 N/cm^2 (Brown & Potvin, 2007), L_m = muscle force/length coefficient determined from anatomical model and equations presented in McGill and Norman (1986), F_p = passive elastic force in the muscle, determined within anatomical model via the equations presented in McGill et al. (1986), Gain = multiplier used to force the EMG determined muscle moment to be equal to the residual moment obtained from the link segment (Q-Back) analysis.

4.3.4.5 Calculations of Trunk Rotational Stiffness (Stability)

Stability about the L4/L5 joint was quantified as the second derivative of the potential energy function describing the trunk system (Bergmark, 1989). The trunk was modeled as a single degree of freedom inverted pendulum (Figure 4-2) and as such, stability was quantified as the rotational stiffness of the system (Stokes & Gardner-Morse, 2003). The system is considered stable when the rotational stiffness is greater than zero. For simplicity, the flexor and extensor musculature represented in Figure 4-2 have been lumped and represented as single equivalents.

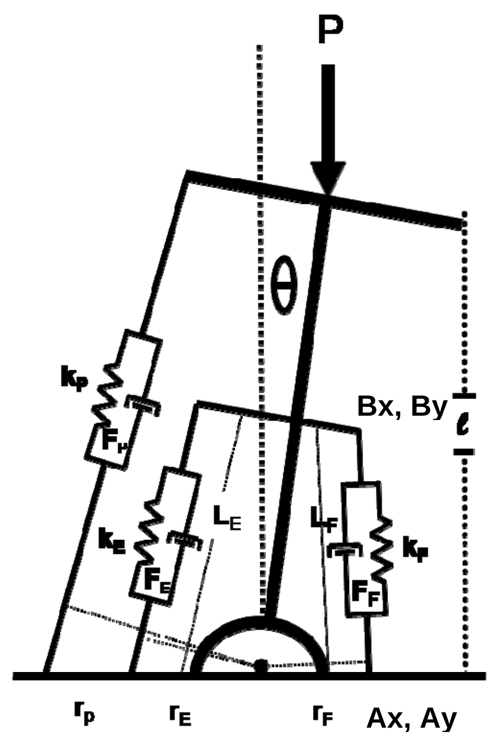

Figure 4-2 Inverted pendulum model used to quantify the effect of the PLAD on trunk (L4/L5) potential energy, equilibrium, and stiffness (stability). Muscle and PLAD stiffness are represented as k_E, k_F, and k_P. Muscle and PLAD forces are represented as F_E, F_F and F_P. L_E and L_F describe extensor and flexor muscle length. Moment arms for the PLAD and musculature are represented by r_P, r_E, and r_F. Sample muscle origin and insertion co-ordinates are represented by A_X, A_Y, and B_X, B_Y. P and ℓ describe the magnitude and height of the load. Trunk angle is represented by θ.

The stabilizing contributions of each flexor and extensor fascicle were calculated individually. Using the muscle force predictions obtained from the EMG model, as well as the associated fascicle geometry, the rotational stiffness (stability contribution) generated by each fascicle was calculated using the equation developed by Potvin & Brown (2005):

$$S(m)_z = F\left(q\frac{A_X B_X + A_Y B_Y - r_z^2}{\ell} + \frac{qr_z^2}{L}\right) \qquad [2]$$

Where $S(m)_z$ = the rotational stiffness of a muscle about the flexion-extension axis of the L4/L5 joint, F = muscle force, as determined from an EMG-assisted biomechanical model, ℓ = Vector length (3D) of the muscle crossing L4/L5, L = Total 3D length of a muscle (if no node is modeled for a particular muscle, L = ℓ), r_z = functional moment arm of the muscle about the flexion/extension axis, A_X,A_Y = muscle origin coordinates relative to L4/L5, B_X,B_Y = node or insertion coordinates of the muscle with respect to L4/L5, and *q* = proportionality constant used to predict muscle stiffness based on muscle force and length (set to 10, from Crisco & Panjabi (1991)).

For trials that featured assistive support from the PLAD, the stabilizing potential of the device was modeled as that of a spring with stiffness (k) acting at distance (r_z). As such, the rotational stiffness of the PLAD was calculated using the following equation (Granata & Orishimo, 2001):

$$S(p)_z = kr_z^{\,2} \qquad [3]$$

Where S(p)z = the rotational stiffness of the PLAD device about the flexion-extension axis of the L4/L5 joint, *k* = the instantaneous, or short range stiffness of the PLAD elements for each trial, and r_z = functional moment arm of the PLAD elements about the flexion extension axis.

From here, the rotational stiffness provided by the musculature and the PLAD were combined to quantify the stability of the trunk. Knowing the contributions of the active musculature and the PLAD, the net stability of the trunk was then calculated for each condition using:

$$\frac{d^2V}{d\theta_z^2} \Rightarrow S_z = S(m)_z + S(p)_z - PL \qquad [4]$$

Where S_z = the rotational stiffness (stability) of the trunk system about the z axis, $S(m)_z$ = rotational stiffness provided by the trunk musculature, $S(p)_z$ = rotational stiffness provided by the PLAD system, and PL = work performed on the external load P (upper body mass) acting at a height L above the L4/L5 joint (Nm).

4.3.5 Statistical Analyses

4.3.5.1 Repeated Measures ANOVA

The effect of flexion and PLAD element stiffness on trunk stiffness (S_z) was contrasted using a two-factor, 5 x 6 repeated measures ANOVA. Significant main effects and interactions (α <0.05) were assessed using paired t-tests and analyzed post-hoc using a procedure known as false detection rate (FDR). Research has suggested that the FDR approach is better suited for multiple contrast tests, compared to other overly conservative techniques, such as the Bonferroni correction procedure (Benjamini and Hochberg, 1995). The FDR technique attains a higher level of power by controlling the proportion of errors that exist among only the tests that had a null hypothesis rejection. All statistical analyses were performed using SPSS 11.0 for Windows (SPSS Corporation, Chicago, IL). Descriptive statistics pertaining to the relative effectiveness of the PLAD system were also calculated in order to further illustrate the effect of the PLAD device on trunk stability.

4.4 Results

4.4.1 PLAD Element testing

Data obtained from a sample stress/strain trial are presented in Figure 4-3. From these data it was apparent that the PLAD elements exhibited the behaviour of non-linear "softening" springs, and as such, became increasingly compliant as they were stretched (smaller increase in force for equivalent change in length as the element was stretched further). The polynomial curve fits obtained for the element tests are presented in Table 4-1. The non-linear elastic behaviour of the elements was consistent across the repeated trials, as evidenced by the high proportion of variance (~99%) accounted for by the polynomial equations.

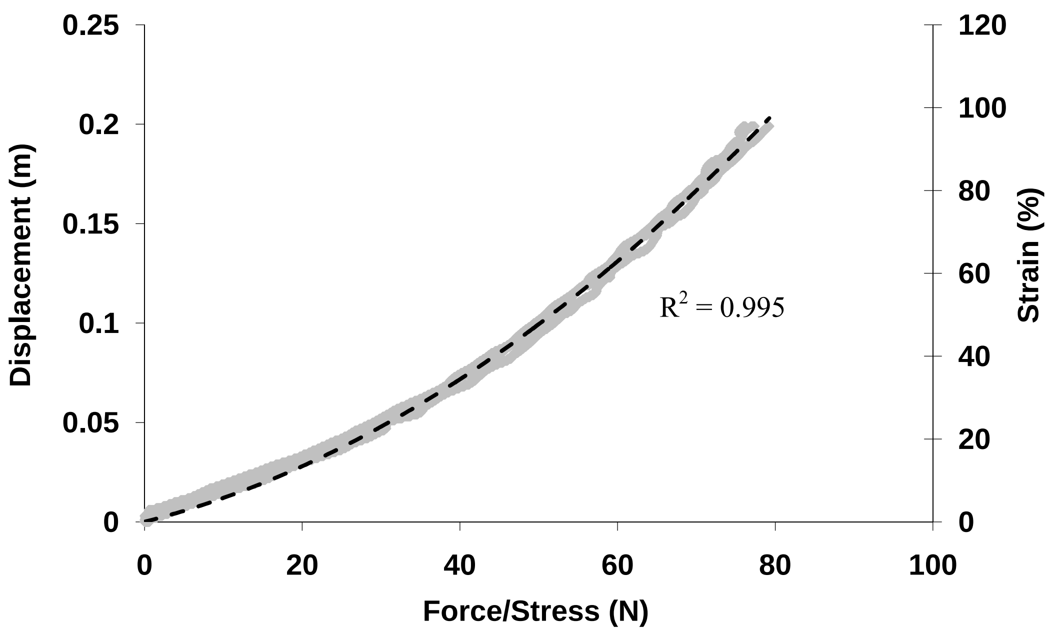

Figure 4-3 Sample trial data (grey) from PLAD element stiffness testing and line of best fit (**black**). The non-linear elastic behavior of the PLAD elements was reflective of that of a "softening" spring.

Table 4-1 Polynomial equations used to estimate PLAD element stiffness within trials based on measures of PLAD element force.

Element	Polynomial Fit – Element Stiffness	R^2
K1	y = 1/(0.0001x + 0.0014)	0.9991
K2	y = 1/(0.00004x + 0.001)	0.998
K3	y = 1/(0.00002x + 0.0007)	0.9991
K4	y = 1/(0.000018x + 0.0004)	0.9976
K5	y = 1/(0.000008x + 0.0004)	0.9962

4.4.2 Estimates of PLAD Rotational Stiffness

Moment arm (r_z) and element stiffness (k) data were used to estimate the rotational stiffness of the PLAD (Figure 4-4). In contrast to increases in passive moment, the rotational stiffness of the PLAD decreased with greater trunk flexion. This trend was consistent for each type of elastic element used within the PLAD. When averaged and compared relative to the 15 degree posture, the rotational stiffness of the PLAD decreased by 39.68 (5.23) %, 62.14 (3.43) %, 74.93 (2.65) %, and 83.18 (1.88) % across the 30, 45, 60, and 75 degree conditions, respectively.

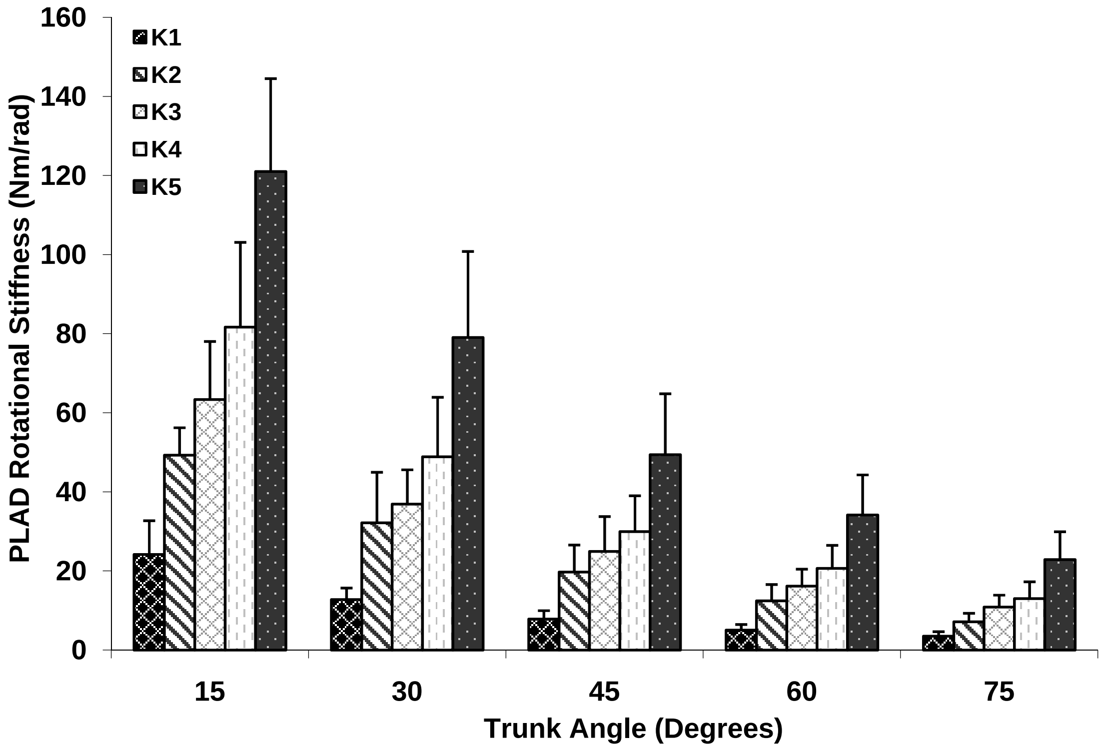

Figure 4-4 Mean (SD) rotational stiffness (Nm/rad) provided by the PLAD contrasted across levels of PLAD element stiffness and trunk flexion angle.

4.4.3 Effect of the PLAD on Trunk Stiffness

Predicted levels of trunk stiffness are presented for each trial condition in Figure 4-5. For the experimental condition of Angle, and the Stiffness x Angle interaction, the assumption of sphericity could not be met based on the results of Mauchly's W statistic. To correct the violation, the degrees of freedom (df) for these factors were adjusted using the Greenhouse-Geisser correction technique. Once corrected, the results of the repeated measures ANOVA indicated significant main effects for Angle, $\underline{F}$ (1.88, 15.103) = 39.722, p = 0.0001 and Stiffness, $\underline{F}$ (5, 40) = 14.591, p = 0.0001, reflective of an increase in trunk stiffness with trunk flexion and a decrease in trunk stiffness as a consequence of using the PLAD. A significant interaction was also identified between the two variables $\underline{F}$ (4.546, 36.608) = 9.729, p = 0.0001. The effects of the interaction are illustrated in Figure 4-5; the net stiffness or stability of the trunk system was increasingly compromised with an increase in trunk flexion and PLAD support.

Using the FDR algorithm and a priori α level = 0.05, post hoc contrasts were deemed significant when $p \leq 0.018$ and were subsequently performed (paired t-tests) to quantify the varying effectiveness of the PLAD across postural conditions. When flexed 15 degrees, the rotational stiffness provided by the stiffest PLAD configuration (K5) significantly increased the overall stability of the trunk (p = 0.001). This equated to a 44.43 Nm/rad (44.32%) increase in trunk stiffness when compared to the No-PLAD condition. In the 30 degree condition, no significant effects were identified between the No-PLAD and PLAD conditions. By 45 degrees of trunk flexion, a significant loss of trunk stiffness was observed for every trial using the PLAD (p = 0.0001, p = 0.0001, p = 0.011, p = 0.001, p = 0.0001; K1-K5 respectively). This net loss in trunk stiffness ranged from 72.56 Nm/rad (-21.65%) to 90.93 Nm/rad (-27.13%). At 60 degrees of flexion, trunk stability was significantly reduced (p = 0.003, p = 0.001, p = 0.003, p = 0.001, p = 0.0001; K1-K5) by magnitudes ranging from 97.47 Nm/rad (-24.31%) to 135.14 Nm/rad (-33.71%) across the five PLAD conditions, respectively. When flexed to 75 degrees, model-based estimates of trunk stability were significantly lower in the PLAD trials compared to the No-PLAD trials (p = 0.0001, p = 0.002, p = 0.0001, p = 0.0001, p = 0.0001; K1-K5), with reductions ranging from 145.82 Nm/rad (-32.98%) to 204.68 Nm/rad (-46.30%) across PLAD conditions.

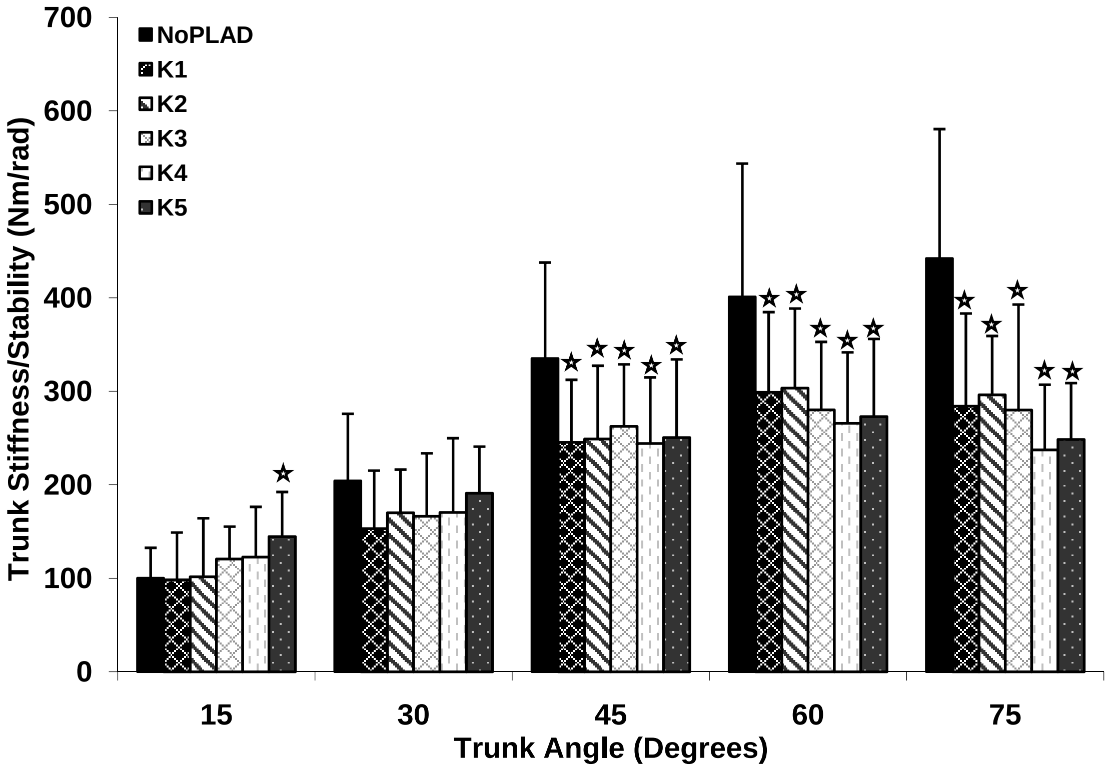

Figure 4-5 The effect of the PLAD on trunk stiffness (stability) across flexed trunk postures. Significant differences between PLAD and No-PLAD conditions are indicated by ☆.

4.5 Discussion

The objective of this study was to quantify changes in active trunk stiffness associated with use of the PLAD. As such, trunk stiffness was quantified across a series of flexed postures through use of an EMG-assisted model and an equation that predicted muscle-based contributions to joint stability (Potvin and Brown, 2005). The effect of the PLAD on active trunk stiffness was documented across this range of flexed trunk postures; multiple levels of PLAD-assistive support were also evaluated. In the No-PLAD condition, trunk stiffness was increased significantly as subjects flexed to greater levels of trunk flexion. This is in accordance with Gardner-Morse & Stokes (2001) and Brown & Potvin (2005) who have documented that trunk stiffness concomitantly increases with extensor muscle activation and/or extensor moment. However, the effect of the PLAD on active stiffness changed significantly over the range of flexed postures tested; this pattern remained invariant across different levels of PLAD assistive support.

At 15 degrees of flexion, there was a general trend across increasing levels of PLAD stiffness which suggested that wearing the PLAD increased the overall stability of the trunk. Accordingly, relative to the No-PLAD condition, a significant increase in the estimated rotational stiffness of the trunk was identified when the stiffest (K5) PLAD element was used. These general findings agree with previous research (McGill et al., 1990; Cholewicki et al., 1999; Ivancic et al., 2002), that suggests assistive devices (i.e., lifting belts and lumbosacral orthoses) add passive stiffness to the trunk during maintenance of upright postures. The rotational stiffness provided by the lumbosacral orthotic used in Cholewicki et al. (2004) was estimated to be 19.7 Nm/rad when the trunk was flexed to 15 degrees. This is comparable to the rotational stiffness provided by the K1 PLAD condition at 15 degrees of flexion (Figure 4-4). However, in the remaining PLAD element stiffness conditions, the rotational stiffness provided by the PLAD was quantified to be two to five times higher than that associated with the lumbosacral orthotic (Figure 4-4) tested in Cholewicki et al. (2004). As such, it is reasonable to assume that the higher levels of PLAD stiffness did increase the overall stiffness of the trunk, as indicated by the significant increase in trunk stiffness observed in the K5 PLAD condition.

While the results suggest that the PLAD can augment the overall stiffness of the trunk in upright postures (up to 15 degrees of flexion), this observation did not continue with increased trunk flexion. Starting at 45 degrees of flexion, use of the PLAD significantly reduced the overall stiffness of the trunk; this trend was evident with every level of PLAD stiffness. Collapsed across levels of trunk flexion and PLAD stiffness, the relative decrease in trunk stiffness ranged 21 – 46%. These results suggest that the overall stability of the trunk is compromised when the PLAD is worn in flexed postures. As such, in the absence of adequate reflex response, a greater perturbation or deflection would be expected if an unexpected force suddenly acted on the trunk during flexed postures featuring use of the PLAD.

This predicted loss in trunk stiffness is directly attributable to the inherent link between muscle force and short-range stiffness. Empirical research has illustrated that muscle stiffness is linearly proportional to muscle force and/or joint moment (Hunter & Kearney, 1982; Granata et al., 2002). Within the trunk stability literature, this proportionality is frequently expressed as the constant "q" (Bergmark, 1989; Lee et al., 2006; Brown & McGill, 2005). Using this constant, muscle stiffness (k) is estimated, as well as an estimate of muscle length (L); yielding $k = qF/L$. As muscle stiffness is dependent on muscle force, any decrease in muscle force reduces the estimated short-range stiffness of the muscle and its contribution to the active trunk stiffness of the trunk. The assistive moment created by the PLAD has been shown to cause significant decreases in the EMG activation levels of the trunk extensor muscles, namely the TES and LES (Abdoli et al., 2006). Due to the functional moment arms and physiological cross sectional areas of these muscle groups, they have been quantified to have the greatest potential for generating rotational stiffness about the flexion-extension axis of L4/L5 (Brown & Potvin, 2007). As documented in Figure 4-4, PLAD rotational stiffness (kr^2) decreases substantially with trunk flexion. This is due to two factors, that being the "softening" behavior of the current PLAD elements (i.e., "k" decreases with increased stretch – Figure 4-3) as well as the decrease in PLAD moment arm that occurs as the trunk flexes forward. It should be noted that the moment arms of the trunk extensors also decrease with trunk flexion. However, in terms of stiffness, muscles exhibit opposite behavior to what was observed with the PLAD elements; specifically: in the absence of external support, trunk flexion leads to greater joint moment; this requires greater muscle force to remain in equilibrium, and ultimately results in higher muscle and trunk stiffness.

These results suggest that an inherent tradeoff exists in terms of PLAD moment and rotational stiffness; the moment created by the PLAD reduces the activity of the trunk extensor musculature. However, while this is beneficial in terms of reducing the physical demands placed on the musculature and the intervertebral joint, the active stiffness (stability) of the trunk is compromised as a result of these decreases in trunk extensor force, specifically those associated with the LES and TES. Based on this finding, redesign of the PLAD elements should be considered. If possible, the elements should be replaced with linear springs, or non-linear stiffening springs if they can be created for this type of application. In addition, an attempt should be made to redesign the PLAD moment arm so that it increases or stays constant across a range of flexed trunk postures. If trunk stability is a critical design criterion, these redesigns should be undertaken in an attempt to minimize the net loss in active trunk stiffness that comes as a consequence of the assistive nature of the PLAD system.

This study was subject to several limitations and assumptions that should be addressed. The largest assumption associated with the study is the postulation that trunk stiffness can be predicted through static postures and monitoring of EMG signals. Quick release or perturbation studies are being used more frequently to assess trunk stability (Cholewicki et al., 2000; Lee et al., 2006; Moorhouse & Granata, 2007). These methods facilitate quantification of the behavior of the trunk post perturbation, and they measure the effective stiffness of the trunk and the contribution of reflexive control, which has recently been shown to play a greater role than thought in stabilization of the spine (Moorhouse & Granata, 2007). However, as suggested in Ivancic (2002), EMG-based estimates of trunk stiffness are highly correlated (R^2 = 0.71) with results describing the post-perturbation behavior of the trunk. Therefore, while systematic error was present in the current study, it was assumed that it had minimal impact on the interpretation of the results, based on the isometric nature of the task and the low activations involved.

Furthermore, errors due to the model used were thought to remain constant across treatment levels, and as such, were partially accounted for through use of a repeated measures design.

The magnitude of the q value used in the study could have confounded the results. This value describes the theoretical relationship between muscle force and stiffness. As such it was a predictive estimate. The rotational stiffness of the PLAD was estimated directly and then summed with the results obtained from the stability model. As such, the selection of a q value too large in magnitude could have produced rotational stiffness values much larger than those quantified with the PLAD, essentially burying the absolute and relative stabilizing effect of the PLAD. To assess the influence of the q value on the results, the data were re-analyzed using a q values that set estimates of stability for the 15 degree, No-PLAD condition to be zero (unstable). As such the active stiffness provided by the musculature was set to the lowest possible theoretical value (Figure 4-6). Even in this scenario, which minimizes the influence of the predicted muscle stiffness contributions to stability, the results clearly indicate the inherent loss of active trunk stiffness, due use of the PLAD, as the trunk flexes forward.

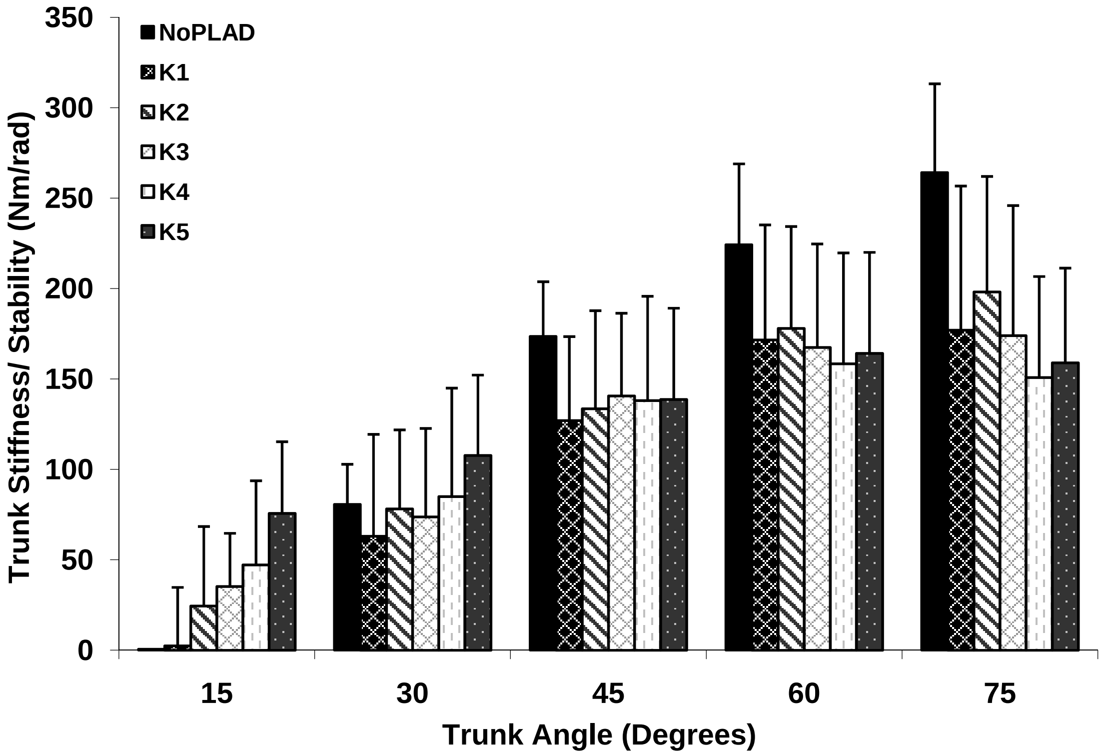

Figure 4-6 The effect of the PLAD on active trunk stiffness when muscle force stiffness contributions are set to their theoretical minimum (min q value).

In summary, the results of the current study present an upper boundary, or worst case account, detailing the effect of the PLAD on trunk stiffness and stability. Spinal stability increases as a function of joint compressive load and muscle activation/co-activation (Stokes & Gardner-Morse, 2003; Vera-Garcia et al., 2006). With this in mind, active trunk stiffness and the effects of the PLAD system were quantified through an analysis of static flexed postures that were performed in the absence of any external load and with relatively low levels of muscle activation. With increased task demands and muscle activation levels, the relative effect of the PLAD device will become diminished. However, the results do provide an understanding of the trade-off that currently exists in terms of assistive support provided by the PLAD and subsequent decreases in active trunk stiffness. Based on the results, recommended design changes have been made, particularly related to developing a design that attempt to preserve the mechanical advantage (i.e., moment arm) of the PLAD throughout a range of trunk flexion. Future research should also focus on documenting the stabilizing effects of the PLAD about the lateral bend and twist axes, and at multiple vertebral levels. Finally, a comparison of the stabilizing potential of the PLAD within the female population should also be conducted, as evidence suggests inherent differences between the genders in terms of active musculoskeletal stiffness (Granata et al., 2002). With these research goals in mind, further knowledge will be gained regarding the costs and benefits associated with use of the PLAD as an ergonomic device and its suitability for use in the workplace.

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Chapter 5
General Discussion

5.1 Introduction

The results of epidemiological research suggest that occupational risk factors, namely physical workplace exposures, account for 30-40% of reported LBP episodes and injuries (Norman et al., 1998; Punnet et al., 2005). Among these, peak and cumulative spine loads (compression and shear), workplace postures, and lifting kinematics have been identified as independent risk factors for LBP (Chaffin and Park, 1973; Kumar; 1990; Marras et al., 1993; Norman et al, 1998). Researchers also suggest that occupationally-related events that cause sudden or rapid movements can be predisposing factors to LBP as well, even when the applied loads are small (Davis et al., 2000). Events such as these have brought forth the notion that injuries leading to LBP can in fact result from joint instability, which is reflective of an inability of the neuromuscular system to adequately stiffen the joints prior to a perturbation (McGill, 2002). With this in mind, it is paramount that these kinetic and kinematic risk factors be considered when any type of assistive device or ergonomic aid is created or recommended, with the ultimate intent of reducing the physical low-back demands associated with an occupational task.

Over the last five years, researchers within our lab have been developing an ergonomic lifting aid known as the Personal Lift Assist Device (PLAD), which is designed to reduce the physical demands and biomechanical loads placed on the tissues of the low back. We have continually assessed the PLAD's effectiveness using kinetic and kinematic outcome measures (Abdoli et al., 2006, 2007, 2008; Lotz et al., 2007) and have done so in an iterative process, refining and redesigning the device in response to our research findings. The focus of this thesis was to further our understanding of the effectiveness of the device and contribute to the existing research that has highlighted the potential benefits associated with this use. Presented here is a summary of findings, followed by a section focusing on limitations of the current research and future research directions.

5.2 Overview of Findings

The purpose of this work was to quantify the effect of the PLAD on select kinematic and kinetic lifting variables that have been indentified as potential risk factors for low back pain (LBP), namely: spinal loads (compression and shear), lifting postures, and trunk stiffness (stability). To do so, three investigations were undertaken in an attempt to further our understanding of the adaptations that occur as a consequence of wearing the PLAD system. Results of these studies are summarized below.

The first paper, entitled ***"Reducing Spinal Loads through Use of a Personal Lift Assist Device (PLAD): Effects of Trunk Flexion and Level of PLAD Support",*** sought to quantify the effects of the PLAD on intervertebral compression and shear loads at the L4/L5 joint. To do so, 30 experimental conditions were tested, varied by trunk flexion angle and level of PLAD assistive support (including a No-PLAD condition). Postural data, PLAD forces and moments, and normalized EMG data from seven trunk muscle sites were used as inputs into an EMG-assisted biomechanical model to yield estimates of L4/L5 compression and shear force. Use of the PLAD significantly reduced joint compression by magnitudes of 191-646 N across the range of trunk flexion; in relative terms, this equated to a 9-24% decrease in joint compressive force. These results compared favorably to those obtained from the mathematical proof developed in Abdoli et al. (2007). Significant changes in shear force were also found, although this effect was much more varied across conditions. Furthermore, it was apparent that the PLAD created a much larger posterior shear force than what was described in Abdoli et al. (2007). The results of the study illustrate the potential benefits of the PLAD as a means of reducing peak spine and cumulative spine loads incurred during occupational tasks requiring lifting and static flexed postures.

The second paper, entitled ***"The Impact of a Personal Lift Assist Device (PLAD) on Lifting Kinematics and Co-ordination",*** focused on documenting adaptations in lifting posture, lift kinematics, and co-ordination due to use of the PLAD. Over two randomized testing sessions, subjects were required to complete a 5-minute repetitive lifting task with and without use of the PLAD; subjects were encouraged to lift with a self-selected technique. Kinematic data describing motion about the lumbar spine, hip, and knee were used to quantify lifting posture, lifting velocities, and segmental co-ordination through estimation of relative phase angles. The results of the study suggest that use of the PLAD caused users to lift with significantly less lumbar flexion (~9 degrees) and significantly greater flexion (~5 degrees) about the hip. Significant decreases were also observed in terms of the relative phase angle that existed between the lumbar-hip and lumbar-knee segments. This might be due to the assistive nature of the PLAD and the ensuing decrease in the physical demands placed on the erector spinae and hamstring musculature. The results of the study further illustrated the potential of the PLAD as a viable ergonomic control solution; however, the differences observed in co-ordination-based measures reflected a distinct change in lifting technique. Further research is needed to understand the long-term effects related to use of the PLAD.

The final paper, entitled ***"The Effect of a Personal Lift Assist Device (PLAD) on Active Trunk Stiffness: Implications for Spine Stability",*** attempted to assess whether use of the PLAD would compromise the stability of the trunk through a reduction in active trunk stiffness. Subjects were required to assume a series of static, symmetrical flexed postures varied by trunk flexion magnitude. Within each posture, a No-PLAD condition was tested, as were the effects of five different levels of PLAD rotational stiffness. Normalized electromyographic (EMG) data were quantified across seven muscles on the right side of the body. EMG, postural, and PLAD stiffness data were input into EMG-assisted stability model of the trunk. Around neutral trunk posture, up to 15 degrees of trunk flexion, the results suggested that the PLAD increased the overall stiffness of the trunk. However, use of the PLAD significantly compromised the active stiffness of the trunk as flexion increased (21-46% decrease). This effect was consistent across all PLAD conditions. Based on results, use of the PLAD with its current design might place workers at an increased risk of injury in events where sudden loads or unexpected perturbations were applied to the trunk. Future research is recommended, as are design changes regarding the function of the PLAD and its components.

Collectively, the results of the studies presented in this documented could be argued as mixed; specifically in terms of supporting use of the PLAD system as an ergonomic assistive device. While studies one and two do lend support to the notion that the device could serve as a beneficial form of engineering control, the results of the third study imply that use of the PLAD could increase the potential for injury, particularly in situations where sudden or unexpected loads are applied to the trunk. With this in mind, it is important to note that the first two studies primarily focused on physical attributes (i.e., forces, postures, and kinematics) that have been recognized, or "accepted" within the literature as work place variables that pose injury risk on operators. These trends have been repeatedly documented through a combination of epidemiological data, and results from in-vitro tissue tests (Marras et al., 1993; Norman et al., 1998; Callaghan et al., 2001; Gunning et al., 2001).

In contrast, the concept of injury to due mechanical instability, or "buckling" is relatively new in concept. While back pain/injury research related to this area has certainly grown in recent years, the fact remains this is potential injury mechanism has yet to be formally indentified as a risk factor for injury, and a causal relationship has yet to be established based on epidemiological data. In fact, to the author's knowledge, only one study has ever formally captured, or documented a "buckling" event to occur during any type of lifting task. In this particular case, the participant in the study was an elite power lifter attempting to lift a load that in well exceeded any load that would be typical to an industrial manual material handling tasks (McGill, 2002). For these reasons, while the results of the third study do offer insight to the overall effect of the PLAD on trunk stiffness, the results do not carry as much importance, or credence as those found in the first and second study. Furthermore, while a decrease in trunk stiffness was quantified to happen as a consequence of wearing the PLAD while assuming a static flexed posture, the predicted stiffness values do not approach a critical buckling load, in fact the estimated buckling loads remain essentially an order of magnitude greater than what would normally be expected in an industrial setting that is currently following established ergonomic weight limits for manual material handling (e.g., NIOSH MWL of 27 kg). Therefore, while the results of study three do identify potential negative attributes associated with use of the PLAD, if a relative weighting was assigned across studies, the results of study three should not be viewed as equivalent to those found in the first two studies. Despite this, the results of study three and the remaining studies all provide insight towards potential redesign(s) of the PLAD and its mechanics, and may serve as criteria, or objective functions to be met in future design iterations, as discussed below.

5.3 Limitations and Future Directions

Given that the PLAD and its current design is very much a work in progress, numerous research questions are still outstanding and must be addressed. While the results presented within this thesis have illustrated the potential of the PLAD as a viable form of ergonomic control, they have also suggested the need for further research and development. Furthermore, the results are also subject to limitations that must also be recognized. An itemized list of research limitations and future research directions is listed below, organized by the research questions addressed in this thesis.

5.3.1 Effect of the PLAD on Spinal Loads

The results of this study indicated that the PLAD significantly reduces spinal loads incurred on the L4/L5 joint during lifting. However, this analysis was performed using symmetrical, static postures and was conducted using male participants. As subjects were required to assume identical postures in both PLAD and No-PLAD conditions, the results of this study reflect the biomechanical benefits solely attributable to the mechanical advantage of the PLAD elements. These data were determined through use of an EMG-assisted biomechanical model. EMG-assisted models are complex, and utilize numerous assumptions related to the EMG-force modeling, link-segment modeling, anthropometry, as well as muscle physiological and geometric data. Finally, a correction factor must also be used in order to equate the EMG-based estimates of force and moment to those determined via a link-segment biomechanical analysis. As such, the kinetic data derived from such an analysis is predicted, as opposed to directly measured and is systematic error is inherent to use of this methodology.

With this in mind, the work presented in this thesis attempted to control for this error as much as possible, primarily through use of a repeated measures (within subject) research design, and by using a set of skeletal model co-ordinates and modeling assumptions that have been argued to be most representative of the anatomy of the low-back, as these data were derived from cadaveric preparations and in-vivo imaging techniques (McGill, et al., 1986; Brown and Potvin, 2007). It must be noted that intervertebral disc pressures are very difficult to measure; the currently available method for directly quantifying these data requires insertion of an pressure transducer directly into the nucleus pulposus of a disc, a surgical procedure that is inherently dangerous to the participant, and limiting in terms of the motions and tasks the participant can perform while the needle is inserted. Because of this, the use of EMG-assisted modeling has been a common practice of researchers interested in low back kinetics, and in order to maintain the highest validity in these measures, several researchers have refined and developed the required assumptions and procedures over the years. While these methodological steps were utilized within this study, their limitations must remain recognized within the framework of predicting biomechanical joint loads.

As indicated in this thesis, use of the PLAD causes changes to occur in lifting kinematics as well; the kinetic benefits associated with these changes in lifting posture remain to be quantified. Therefore, any future work assessing the effect of the PLAD on intervertebral joint loading should involve a methodology that involves lifting tasks (as opposed to equivalent static postures) and subjects should be provided adequate practice with the PLAD prior to testing. Furthermore, research should attempt to quantify the effectiveness of the PLAD at reducing spinal loads associated with asymmetrical lifting tasks and postures. Finally, since the PLAD elements cross all of the lumbar vertebrae, future work should focus on quantifying the effect of the PLAD across the entire lumbar spine.

5.3.2 Effect of the PLAD on Lifting Kinematics and Co-ordination

The results of this study indicated that the PLAD influences users to adopt a lifting posture that is composed of less trunk flexion and greater flexion at the hips. The study also suggested that differences in co-ordination occur with use of the PLAD. The findings of this study are limited, due to the fact that only one load and lifting height were tested and that the task was symmetric, a trait not easily found in industrial settings. Furthermore, kinematic data were somewhat limited in that they derived for the knee, hip, and lumbar spine only. In addition, the dependent variables used within the study were general parameters of curves, and in many instances, only described a discrete component or period of the lifting task. While an attempt was made to account for this by way of assessing both the approach, and lifting components of the task, again these data serve as relatively crude, discrete parameters that can be used to describe a curve, or waveform. To build on these findings, more advanced analysis techniques, such as principal component, or functional analyses of variance (fANOVA) should be considered, as recent evidence has illustrated that these methodologies provide further insight simply beyond single curve parameters, such as mean values, peaks, etc (Godwin, et al., in press). Future research should also attempt to identify if the results of the current study are present in asymmetrical tasks varied by task, load, and frequency. In particular, future research should be dedicated to further understanding the co-ordination-based changes witnessed in the current study to assess whether they are indicative of a change in motor control strategy due to the assistive support of the device.

5.3.3 Effect of the PLAD on Active Trunk Stiffness

The results of this study suggested that active trunk stiffness is compromised due to use of the PLAD. The findings of this study were limited particularly due to the methodology used because trunk stiffness was predicted as opposed to directly measured. PLAD element stiffness was predicted as well, as opposed to directly measured within each experiment. Furthermore, the PLAD elements were modeled to be rigidly attached to the trunk; as such, the compliance of the shoulder padding was ignored. Despite this, the results suggested the need for redesign of the PLAD, most importantly the PLAD elastic elements. The current PLAD elements become more compliant as they are stretched, and as such, the rotational stiffness of the PLAD decreases as the trunk flexes forward. In contrast, the short-range stiffness of skeletal muscle increases proportionately with muscle force, creating an increase in active trunk stiffness as the trunk flexes forward. The results suggest that, as the trunk flexes forward, the loss in trunk stiffness due to decreased muscle activity is not equally matched by the PLAD, thus decreasing the stiffness of the spine, and its margin of safety, or ability to absorb perturbation energy from a sudden, or unexpected load. Future research should attempt to recreate these findings, if possible, through use of a quick-release perturbation study. Furthermore, an alternate PLAD element should be tested; in particular, one that exhibits non-linear (stiffening) behavior and more closely represents the stiffness characteristics of skeletal muscle.

Finally, as documented within studies one and three, the PLAD moment arm decreases as trunk flexion increases. This change in moment arm, combined with the observed decrease in element stiffness, results in a decrease in the amount of assistive trunk stiffness provided by the PLAD as a user assumes a flexed posture at the trunk. These findings serve as a rationale toward future investigations that should attempt to resolve this problem, and primarily work toward refining the current design of the PLAD moment arm. For example, the current PLAD design features one lever arm that projects dorsally from the waist belt, and 2 attachment points that are located at the shoulders, and knees, respectively. The potential for multiple lever arms should be considered, such that multiple attachment nodes or pulleys are located between the shoulder attachments and the main lever arm located on the waist belt. By doing so, the PLAD element line of action will not decrease to as great an extent as in the current configuration, and as rotational stiffness is a function of both element stiffness, as well as moment arm (squared), this design change should translate into an increase in the rotational stiffness provided by the PLAD, especially during high levels of trunk flexion. This type of mechanical design is characteristic of the larger skeletal muscles of the back (e.g. erector spinae) as these muscle groups utilize multiple tendonous attachment points or "nodes" that are located along the spinous processes of vertebrae. These nodes serve as a means of preserving the mechanical advantage of the muscle fascicles, and maintaining their line of action over the intervertebral joints that they span. As the initial concept of the PLAD system was to develop a passive system that provided assistive torque with similar lines of action as the musculature of the low back, the rationale for an investigation that seeks to redesign the mechanical advantage of the PLAD in this way is certainly warranted.

5.4 General Summary

As part of the effort towards reducing the physical demands, inherent injury risks, and subsequent manifestation of musculoskeletal injuries amongst workers required to perform manual materials handling tasks (MMH), and occupational tasks requiring prolonged flexed postures, investigators have attempted to implement and subsequently quantify the effectiveness of various administrative and/or engineering control strategies. In terms of engineering-based controls, numerous potential solutions have been evaluated, ranging from postural/workspace design recommendations to complete mechanization and automation. Additionally, the use of load transfer devices has been suggested as they have been shown to reduce stresses placed on the low back during excessive stooped, twisted postures.

With this is mind, research within our laboratory at Queen's has focused on developing a personal lift assist device (PLAD) for comparable use in manual handling tasks, and occupational tasks requiring prolonged stooped postures. To date, numerous research studies have been undertaken within our laboratory in order to document the effectiveness of the PLAD as a viable ergonomic aid, as well as to evaluate the various redesigns and improvements that have been made along the way. The results of these efforts have quantified the effectiveness of the PLAD at: generating assistive moment (Abdoli et al., 2008), reducing erector spinae EMG amplitude for a given lifting task (Abdoli et al., 2006, Stevenson et al., 2007), and diminishing the manifestation of localized muscle fatigue in the low back musculature over periods of repetitive lifting (Lotz et al., 2008; Godwin et al, in press). Building on this previous research, the results presented in this thesis expand upon our understanding the efficacy of the PLAD system, and its potential as an ergonomic aid. Based on the results of the research within this thesis, it is reasonable to assume that for a given posture, or lifting task, use of the PLAD decreases the biomechanical joint loads experienced in the low back (i.e., L4/L5) and that when used as an intervention, the PLAD influences users to lift with a decrease in lumbar flexion, and an increase in hip flexion, a technique that has been argued to be "safer" in terms of lifting strategy (McGill, 2002). However, the results of the third study suggest potentially negative attributes associated with use of the device, and dangers related to instances where sudden loads or perturbations take place.

For these reasons, it is apparent that future research dedicated to the device should also focus on redesigning the current configuration such that these negative attributes are minimized. Additionally, it must be stressed that, to date, all of the studies focusing on assessing the efficacy of the PLAD have been restricted to a lab setting. As such it is imperative that once all design modifications are finalized, the true effectiveness of the PLAD be assessed by way of a longitudinal field study within an industrial setting. Once completed, the benefits associated with use of the PLAD as an ergonomic aid will be defined and known, and future research regarding the implementation and success associated with use of the PLAD as an ergonomic control solution can be pursued.

5.5 References

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Appendix A: Sample Consent Form

The effect of a Personal Lift Assist Device (PLAD) on muscular fatigue during a repetitive lifting task

Dear Participant,

You are invited to participate in a research study to examine the effectiveness of a personal lift assist device (PLAD) at reducing muscular fatigue. This investigation is being conducted in the Biomechanics Laboratory at Queen's University. We will read through this consent form with you and describe the procedures in detail. You will be given time to read it yourself and are encouraged to ask questions at any time.

Aims and Purposes of the Study:

The goal of this research is to quantify the ability of the PLAD to reduce muscular fatigue during a lifting task. The data will be gathered using a Fastrak® motion sensor system and a Bortec® electromyography (EMG) system. The Fastrak® system detects markers that will be placed on selected skin surfaces. These data will be used to examine changes in the lumbar curvature and lifting technique throughout the session. Data will also be collected from EMG electrodes placed on eight different muscles on the right side of your body. These data will be used to examine muscle activity during lifting. The main objectives of this study are to assess the effectiveness of the PLAD in reducing muscular fatigue during lifting, and to determine if there are any changes in lifting technique.

Study Details:

This is a cross-over design and you will only wear the PLAD during the first or the second session. The session at which the PLAD will be worn will be determined randomly (like flipping a coin). The sessions will last 2 to 3 hours each, with one week between sessions.

Anthropometry and Health Examination: You will be asked to complete a Physical Activity Readiness Questionnaire (PAR-Q) in order to assess your general health prior to a physically demanding task. We would appreciate it if you would inform us of any allergies or skin irritations that you may have experienced in the past (ie. rubbing alcohol, isobutene, propane, or rosin). We will also exclude you if you suffer from low back pain within the last two years. After you have been cleared to take part in the study, we will record your height (m), weight (kg), specific body dimensions, and age.

Warm-up: Exercising muscles that are not often used may result in muscle soreness for 2-5 days after the sessions. To help combat this soreness and to protect against musculoskeletal injury, you will be asked to perform a five-minute warm-up to help protect your muscles and joints.

EMG Electrodes: We will tape eight sets of surface electrodes to muscles on the right side of your body including: the erector spinae at the level of L3, and T9, external oblique, rectus abdominis, biceps femoris (hamstring), and rectus femoris (quadriceps), anterior deltoid, and the gastrocnemius. A reference electrode will be placed on C7. In order for the sensors to stick to your skin, we will first shave a small patch of hair (if necessary) from the sites for electrode placement, using softening foam and a hand-held razor. Rubbing alcohol will be applied against your skin using a cotton pad in order to remove any dead skin cells. After the area has dried, we will spray Quick-Drying Tape Adhesive (Q.D.A) on your skin and secure the electrodes with stretchy fabric-based adhesive tape. We will ask you to bend forward to mark the areas for placement at T9 and L3.

Heart Rate Monitor and RPE Scale: You will be outfitted with a Polar® heart rate monitor to record changes in exercise intensity. If your heart rate approaches 80% of your maximum (220-age), you will be asked to stop the task. We will be recording heart rate every minute during the task. In addition, you will be asked to rate your perceived exertion on a scale from 6 to 20 every minute, where 6 corresponds to "no exertion at all" and 20 corresponds to "maximal exertion".

Fastrak® Sensors: Your right thigh length will be measured in order to determine the centre of gravity. A Fastrak® sensor will be placed over this site. In addition, three other sensors will be placed on the right anterior superior iliac spine (ASIS), at the first lumbar vertebrae (L1), and at the sternum. We will ask you to bend forward so that we can find the landmark for L1. The sensors will be mounted using Q.D.A. (Cramer®) spray followed by stretchy fabric-based adhesive tape.

Maximum Voluntary Contractions: We will ask you to complete three maximal tests for each muscle site so that we can capture maximal EMG activity. For each test, we have set-up a protocol and posture for you to assume in order to elicit the most accurate contraction. By pushing or pulling against an immovable anchor, we will record your contractions. You will be required to hold the contraction for approximately 5 seconds before relaxing. We will provide one minute of rest between each trial and more rest if requested. In addition one maximal back strength test will be performed using a modified Lido® lift, which will allow subjects to stand erect and pull against a strap attached to a load cell. This load will be used to determine the weight of the box, and also fifty percent of that maximum value for the submaximal tests. One maximum voluntary contraction of the erector spinae will be taken at the beginning and end of each testing session.

Submaximal Test Contractions: We will ask you to complete three submaximal tests every five minutes during the dynamic lifting protocol. These tests will be performed using the modified Lido® lift, where you will be secured in an erect standing position and you will pull on an immovable strap using your back muscles. You will be asked to perform a submaximal contraction at 50% of your maximum contraction for five seconds. Five seconds of rest will be given, and this test will be repeated two additional times.

Endurance Test: You will be asked to perform an endurance test using 50% of your maximum contraction calculated based on the maximum contraction using the Lido® lift. The first endurance test will be performed on the first day of testing, and no dynamic lifting will be performed. On the second and third day of testing, you will perform this test at the end of the session. You will be asked to perform a static contraction in an erect standing position secured to the Lido® lift for as long as you can stay above 50% of your maximum. As soon as you drop below 30% of your maximum, the test will be ended.

Wearing PLAD: When you wear the PLAD, you will be fitted by tightening the straps around the shoulders, waist, and knees. We will adjust it until you are comfortable and willing to proceed with the lifting session. Testing with the PLAD or no-PLAD on either day will be determined randomly.

Lifting Protocol: During the lifting protocol you will be required to lift a wooden box with open handles. All the lifts will be executed in the saggital plane, using a free-style lifting posture. A metronome will be used to ensure that you perform the proper number of lifts per minute, at a relatively constant velocity.

You will be taken through the protocol and asked to perform practice trials prior to being outfitted with the equipment to determine your preferred lifting style. In addition, you will be asked to perform a few practice trials (approximately 10 lifts) after the equipment is set up. You will be asked to begin in a relaxed standing posture and pick up the box from the floor and place it on a shelf. The shelf will be adjusted to your waist height. The position of the load with respect to your body will be determined after the practice trials and this position will be marked on the floor to ensure that the load is lifted from the same spot for each lift.

A load of 15% of your erector spinae MVC taken from the Lido® lift will be used. You will be asked to perform six lifting cycles per minute over 45 minutes (270 cycles). One cycle will include lifting the box to the shelf, and lowering the box back to the floor. Every five minutes you will be required to perform three static holding tests for your back muscles at 50% of your MVC. Your heart rate and rate of perceived exertion will be monitored and you may be asked to discontinue the study if you level of exercise intensity becomes too high. You can also choose to end the lifting task at any time during the testing if you feel pain or discomfort. After each day of testing is complete, all the equipment will be removed and De-Hesive (Cramer®) spray will be used to remove all excess tape and residue.

Risks and Benefits of Participation:

This study involves both maximal contractions against an immovable anchor and a series of lifts at a consistent pace. Lifting boxes does increase your risk of low back pain or musculoskeletal injury. To protect you as much as possible from this risk, we will ask you to warm-up your muscles prior to performing maximum voluntary contractions. We have also selected weights that are within NIOSH lifting guidelines for manual materials handling. Your heart rate and perceived exertion will be monitored throughout the study to ensure that you do not tax your cardiovascular system. Exercising muscles that are not used often can leave you with a sore feeling in your muscles for a couple of days after the study. If you have soreness or pain, please call the investigators and they will make a referral to a health care professional.

Sensors will be applied using Q.D.A. (Cramer®) spray followed by stretchy fabric-based adhesive tape. This may cause some skin irritation, which normally disappears shortly after the tape is removed. Applying electrodes means that the skin needs to be abraded using rubbing alcohol and cotton pads in order to remove dead skin cells to improve the EMG signal. This may cause some skin irritations, which normally disappear shortly after electrodes are removed.
In the event that you are injured as a result of participating in the study, we have access to the Athletic Therapists in the Physical Education Centre and have an emergency plan in place.

In terms of the benefits of the study, there are no direct personal benefits expected. You will have an opportunity to contribute to the development of a new on-body personal lift assist device that may aid workers reduce the risks of low back pain.
By signing this consent form, you do not waive your legal rights nor release the investigators from their legal and professional responsibilities.

Confidentiality:

All information obtained during the course of this study is strictly confidential and your anonymity will be protected at all times. Your identity is only recorded at the time of filing the consent forms. You will be assigned a study number which will link your information to this file. All data recorded in computer files will be locked and only the principal researcher and research assistants will be granted access. In all cases of publication, only summary data are used and this is done in such a way that no individual can be identified.

We are interested in collecting video and digital photographs for use in future presentations and publications. These photographs are not needed for data analyses, merely to demonstrate the PLAD and to demonstrate any changes in lifting technique. If you are willing to have your photograph taken, we will ask you to sign the section at the bottom of this form pertaining to this information.

Voluntary Nature of the Study:

As a participant, you are a volunteer who may withdraw from the study at any time without coercion or penalty. You may withdraw after hearing about the details of the study. You may also withdraw at any point during the study with no penalty. You will receive full financial compensation up to the time of withdrawal. If you choose to withdraw, we will remove all of your data from the database.

Compensation:

You will receive a stipend of $10.00 for each hour you participate in this study. Typically, the entire study will require a total of two hours per session.

Participant Statement and Signature:

I have read and understand the consent form for this study. The purposes, procedures and technical language have been explained to me. I have been given sufficient time to consider the above information and to seek advice if I choose to do so. I have had the opportunity to ask questions which have been answered to my satisfaction. I am voluntarily signing this form. I will receive a copy of this consent form for future reference.

Contacts:

If at any time I have further questions, problems or adverse events, I can contact:

Michael Agnew School of Physical and Health Education	at	(613) 533-2658
Christy Lotz School of Physical and Health Education	at	(613) 533-2658
Dr. Joan Stevenson School of Physical and Health Education	at	(613) 533-6288
Dr. Janice Deakin School of Physical and Health Education	at	(613) 533-6601

If you have any questions regarding your rights as a research participant, you can contact:

Dr. Albert Clark Research Ethics Board, Chair	at	(613) 533-6081

What Does My Signature Mean?:

By signing on the following page, I am indicating that:

- I have read the letter of information
- I am aware that the purpose of the study to assess a personal lifting assist device (PLAD)
- I realize I can withdraw at any time without penalty or coercion
- I can contact any of the people identified in this letter if I have questions, concerns, or complaints
- I realize that my data will be kept confidential. Only if I sign a second time below will additional photos or video be taken for possible use in presentations or publications
- By signing this consent form, I do not waive my legal rights nor release the investigator(s) and sponsors from their legal and professional responsibilities.

Signature Page (Sign two times, one for yourself and one for the investigator):

______________________________________ ___________________

Signature of Participant Date

Statement of Investigator:

I have carefully explained the nature of the above research study. I certify that, to the best of my knowledge, the participant understands clearly the nature of the study and demands, benefits, and risks involved to participants in this study.

_____________________________ __________________

Signature of Investigator Date

Consent for Photographs:

By signing below, I am indicating my willingness to be photographed for presentations or publications. I realize my identity will be blocked from view in these images.

______________________________________ ___________________

Signature of Participant Date

Appendix B: PAR-Q Questionnaire

Physical Activity Readiness
Questionnaire - PAR-Q
(revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ____

SIGNATURE ____ DATE ________________________

SIGNATURE OF PARENT ________________________________ WITNESS ________________________
or GUARDIAN (for participants under the age of majority)

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

 Supported by: Health Canada Santé Canada

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