ABSTRACT

KELAHER, DANIEL PAUL. Effects of Trunk Extensor Muscle Fatigue on Trunk Proprioception and Biomechanics. (Under the direction of Dr. Gary A. Mirka.)

Localized muscle fatigue of the low back musculature is common in workers who perform repetitive lifting and manual materials handling tasks. A significant amount of research has been performed on the basic physiology and biomechanics of localized muscle fatigue during heavy physical exertions. Far fewer studies have focused on localized muscle fatigue of gradual onset which is more consistent with that seen in occupational environments. Also, none of this previous research has developed a model that evaluates the important interactions of the various biomechanical responses. The current research considered the effects of gradual onset trunk extensor muscle fatigue on the interactive effects of trunk proprioception and biomechanics.

Localized muscle fatigue was generated in six human subjects as they performed sets of 30 sagittally-symmetric isokinetic trunk extension exertions wherein the required extension moments were gradually increased as the experiment progressed. Interspersed among these sets of fatiguing exertions were a series of testing "modules" that evaluated various dimensions of the response to the fatiguing exertions. These modules included a trunk repositioning module, a trunk extension force replication module, a trunk extension reaction time module, a three dimensional lifting task module, an EMG-based, sagittally symmetric trunk extension module and an EMG-based sagittally asymmetric trunk extension module. These six modules were performed before and after the fatiguing exertions to quantify the fatigue-induced changes in each of these measures independently. Further, each

of these tasks was performed multiple times in each module to gain an appreciation for the variability of the response measure. (For comparison, a group of three anthropometry-matched control subjects performed the six modules without performing the fatiguing exertions.) Finally, a correlation analysis of the inter-relationships between these responses was conducted to gain a more complete understanding of the inter-relationships among these responses. Dependent measures considered in these analyses included measures of average performance ("accuracy") and variability of performance ("precision") while the independent variable was time into the experiment (cumulative fatigue). The results of this correlation analysis were then compared with the expectation of the theoretical model to evaluate the fundamental concepts that the model put forward.

The results for the individual tests agreed with previous research performed on the low back extensor muscle fatigue. In general, the following effects were seen as a function of time into the experiment: trunk extensor muscle onset time increased, trunk extensor NEMG increased, trunk repositioning accuracy decreased, force replication accuracy decreased, the lumbar angles in all three planes during the kinematics test decreased at the start of the lift, and peak angular velocities in all three planes increased in the kinematics test. There were very few significant effects of the time into the experiment on the precision or variability of the various measures. The proprioceptive measures of force replication and trunk angle reproduction in all three planes worsened with fatigue. There was an average increase of 37% in the average absolute error in trunk repositioning test while there was almost a 53% increase in average force replication absolute error. Peak lumbar angular velocities also increased with fatigue 17% above the pre-fatigued values, on average.

The data provides support for a proposed model which includes proprioceptive and biomechanical components linked together. The data strongly supported the kinematic links in the model with a number of significant correlations among the different variables above 0.50. However, the data did not support the kinetic links as there were no significant correlation coefficients above 0.50 or below –0.50. Future research that further integrates proprioception and biomechanical measures is needed. Industrial engineers and ergonomists are recommended to focus on the effects of gradual onset muscle fatigue and the posture and motion parameters seen during repeated trunk exertions in industry.

EFFECTS OF TRUNK EXTENSOR MUSCLE FATIGUE ON TRUNK PROPRIOCEPTION AND BIOMECHANICS

By

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DEDICATION

I would like to dedicate this work to the female Kelahers in my family: my mother, Joan, my wife, Lisa, and my daughters, Tessa, Katie, and Megan. God blessed me with a wonderful mother who raised me and my brothers alone after the death of my father and who always supported all my endeavors. I am lucky enough to have a wife, Lisa, who has always supported my academic work despite how much time it took me away from her. I am now lucky enough to be blessed with three beautiful daughters, Tessa, Katie, and Megan. Even they supported my school work as I worked towards my final defense (“my show and tell” according to Tessa or “my school craft” according to Katie) which was just three days before Megan was born. I truly appreciate all the things these wonderful ladies have done for me, and I want Mom, Lisa, Tessa, Katie, and Megan to know how much I appreciate their love and support. In appreciation of their support, I dedicate this work to them all.

BIOGRAPHY

Dan Kelaher is the youngest of five children born to Mrs. Joan and the Late Mr. John Kelaher. Like his four older brothers, Tom, Tim, Dave, and Chris, Dan was raised in New Providence, New Jersey. There he attended grade school and high school. After high school, he spent two years at Marietta College in Marietta, Ohio then transferred to the Exercise Science Department at the University of Massachusetts at Amherst, where he earned a Bachelor of Science degree with a specialization in Biomechanics. In 1992, he enrolled in the Ergonomics Program in the Industrial Engineering Department at North Carolina State University. He completed his Master's Degree in 1996 while living in New York and working for Ergonomic Technologies Corporation, then returned to North Carolina State University to start on his doctorate program. Since beginning his doctoral program, Dan has gotten married to the former Lisa Anne Servidio, has become the father of three wonderful daughters, Tessa, Katie, and Megan. After completing his Doctoral degree he will resume his work as a Human Factors Engineer at IBM in Research Triangle Park, North Carolina where he has worked full-time since 1999.

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Low Back Disorder Epidemiology

Occupational low back disorders impose a serious financial, as well as human toll on industry. Epidemiological studies have found that approximately 50-70% of the population will experience low back pain (LBP) at some point in their lifetime while roughly 10-30% of people suffer from LBP at any given point in time (Andersson et al, 1984). Webster and Snook (1994) found that while only 16% of Worker's Compensation claims are due to low back injuries, they represent one-third of all claims costs. The median cost per case was only $396, but the mean claim cost was more than 20 times higher, at $8321 per case, indicating a highly skewed distribution where relatively smaller number of claims are responsible for a large percentage of the total costs. Moreover, individuals with multiple low back pain episodes account for especially high costs because subsequent episodes account for higher medical and indemnity costs than the initial episodes (Wasiak et al, 2006). These numbers highlight the large number of cases and the wide distribution of the cost of low back disorders. More recent statistics (Bureau of Labor Statistics, 2002) show that overall occupational injuries and illness rates were falling in the late 1990's, but in 2000 the back remained the most common body part affected (24.7% of all cases), while overexertion in lifting represented the largest event/exposure category (15.4% of all cases). Table 1 shows the various low back musculoskeletal injuries that can occur. The most prevalent injuries are muscle strains, while the most costly are disc injuries that may involve nerve root impingement. This table does not include the high number of low back injuries that can not be linked to a specific anatomical location. In reality, many cases do not ever get categorized as clearly as this table represents due to the many factors involved in diagnosing low back disorders.

Table 1: List of common occupational low back disorder categories and associated anatomical components.

Materials	Components	Common Disorders
Bone	Vertebral bodies, spinous processes and transverse processes	Vertebral compression fractures Avulsion fractures Vertebral endplate fracture
Joints	Facet Joints Intervertebral Joints	Arthritis Spondylosis
Cartilage	Intervertebral discs	Disc herniation
Ligament	Supraspinous ligaments and Thoraco-lumbar fascia (among others)	Ligamentous sprain
Muscle	Erector spinae muscles and Abdominal muscles	Muscle strain
Nerve	Spinal cord and nerve roots	Nerve root impingement

Low Back Injury Risk Factors

Both personal and occupational factors can contribute to the development of low back disorders (LBDs) among workers. Personal risk factors such as obesity (Deyo and Bass, 1989; Han et al, 1997), age (Nerlich et al, 1997), smoking (Deyo and Bass, 1989; Battie et al, 1991), and previous injury (Biering-Sorenson et al, 1989; Punnett et al, 1991; Daltroy et al, 1991; Hides et al, 1994) have all been found to increase the risk of low back injury.

Occupational risk factors for LBDs are those factors that engineers and management can alter through changes in the task requirements and/or work methods. These factors are more under the control of the employer since equipment or standard work methods are under the control of the employer, who designs the workplace and establishes the standard work methods. The occupational risk factors for LBDs that are associated with manual material handling jobs are discussed in the following sections.

Load Weight

The most highly discussed cause of occupational LBDs is the weight of objects handled manually. Punnett et al (1991) performed a detailed analysis of workers in an automotive assembly plant. They found that workers who lifted items weighing at least 10 lb. each work cycle (~60 seconds) were roughly twice as likely to request medical attention for low back pain (LBP) than those who did not. Similarly, Kelsey et al (1984) found that workers who were required to lift items of at least 25 lb. were approximately 3.5 times more likely to suffer a prolapsed intervertebral disc. (It is important to note that this was the only study that used a medical diagnosis of a prolapsed disc as the outcome of interest since it is one of the most severe low back injury types. Other studies typically consider a worker a low back injury case if s/he had been to a medical facility for treatment of his/her low back pain.) Marras et al (1993) performed a large cross-sectional study across 14 different industries on the effects of static and dynamic lifting factors on low back injuries. They found that a three-fold increase in the average load lifted yielded a three-fold increase in the probability of the job being considered a high risk job (defined by the authors as a job that experiences greater than 12 new Low Back Injury cases per year per 100 full-time workers). Additionally, they found that the most powerful discriminating factor between the high and low risk groups was the maximum moment of the load about the L5/S1 joint. Herrin et al

(1986) evaluated the biomechanical stress on the lumbar spines of a group of industrial workers using a biomechanical model. They found a low back injury incidence rate 1.5 times higher for jobs that exhibited intervertebral compressive forces between 1000 lb. and 1500 lb. than for jobs that imposed joint compressive forces less than 1000 lb. This study and the results from Chaffin and Park (1973) and Marras et al (1993), highlight the importance of considering the weight of the object in the context of the entire manual material handling task (e.g., the posture assumed when lifting the loads), rather than just the object weight alone.

Muscle Strength Requirements

Low strength and/or low endurance of the low back muscles are personal risk factors for LBDs. However, the strength and endurance demands of a job also include the job-related factors of the load weight and the posture required to lift that load. Chaffin and Park (1973) showed that jobs that exhibited a Lifting Strength Rating (LSR) above 0.50 led to much higher low back injury rates in an electronics manufacturing company. (Note: LSR = Lifting Strength Rating = ratio of load lifted during a given task to the load capable of being lifted by only 2.5% of men, which is based on the weight and the location of the load). This relationship showed that as the strength demands of the task increase the risk of LBD increases. In fact, this relationship was even stronger than the relationship between the weight of the object lifted and low back injuries (LBI). This relationship is shown graphically in Figure 1.

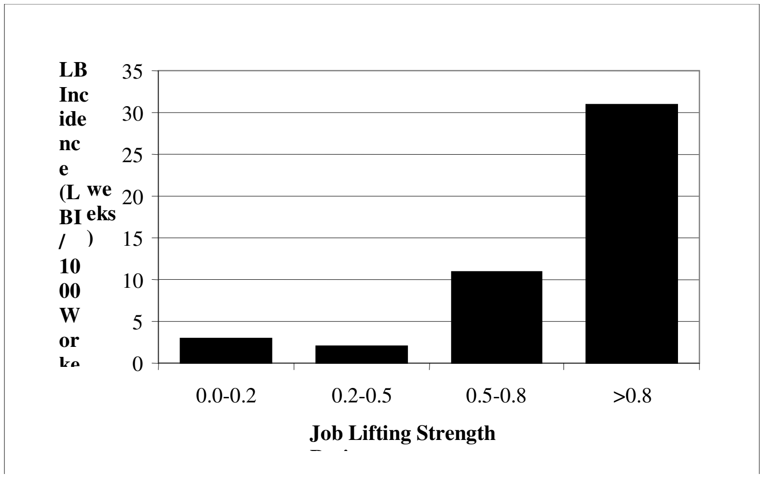

Figure 1: Low back injury incidence rate as a function of LSR (from Chaffin and Park, 1973).

Trunk Posture and Trunk Motion

Awkward trunk postures have been cited as another risk factor for the development of LBDs. In the automotive plant study by Punnett et al (1991), trunk posture was the primary contributor to low back disorders, defined as self-reported back pain for at least one week of length, or three separate instances, over the course of one year. Workers in this study were nearly 5 times more likely to seek medical attention for low back pain if their job required between 20^o and 45^o of forward trunk flexion and nearly 6 times more likely if the job required more than 45^o of trunk flexion. Lateral bending and/or twisting of the trunk also increased the risk of low back injury roughly 6 times above those workers whose jobs did not require them to experience these postures. The combination of mild trunk flexion and trunk twisting yielded the highest injury odds ratio of 7.4, meaning these workers were 7.4 times more likely to suffer low back pain requiring medical attention than the workers who did not experience this combined posture.

The primary focus of the study by Marras et al (1993) was on the dynamic variables of manual material handling jobs. This was the first large-scale field study of its kind to focus on the effects of dynamic trunk variables on low back injuries. They analyzed the angular positions, velocities, and accelerations of the trunk in the three primary planes as workers performed their jobs. Their results agreed with previous research that showed a link between higher LBI incidence rates and awkward posture (e.g., odds ratio of 1.6 for the maximum trunk flexion position). However, their results showed higher odds ratios for the trunk angular velocity parameters (e.g., odds ratio of 3.33 for average angular velocity in the sagittal plane). Fathallah et al (1998b) extended this research by analyzing the interactive nature of these risk factors on low back injury risk. Similar to Punnett et al (1991), Fathallah et al (1998b) highlighted the interactive nature of these risk factors on the risk of low back injury. They found better predictability of a multiple logistic regression model that included an interactive effect of the forward flexion position, twisting angular velocity, and lateral bending velocity than the original model reported in Marras et al (1993) which evaluated each of these variables independently. This shows that there is an important interaction between posture and motion that has not been researched extensively to date.

Temporal Issues

The temporal issues related to manual material handling activities have been especially difficult to quantify and associate with low back injuries. The amount of time a worker spends in a job does not necessarily provide information regarding the risk of low back injury. It can be argued that the longer a worker stays on a job, the higher the risk of injury. However, if a job is stressful for a worker, as s/he gains seniority in the company s/he may try to move to a less stressful position. In fact, job turnover has actually been used as a surveillance tool to help ergonomists identify jobs that are perceived as stressful to workers (Lavender and Marras, 1994).

Exertion duration is another temporal variable that is not well understood, primarily because it is rarely dictated by a job task and is highly variable in nature. The only study that seemed to capture any reliable exertion duration information was the study by Punnett et al (1991). They found that odds ratios for low back injuries increased for any non-neutral posture that was sustained for more than 10% of the cycle time. This information was easy to gather from this subject population, because the nature of an automotive assembly line worker's job is very static in nature and characterized by long durations (20 - 40 seconds) of static posture in each cycle.

The effect of lifting frequency on LBDs has been investigated in previous research. Chaffin and Park (1973) found that workers who lifted very infrequently (less than 50 lifts/day) experienced LBI incidence rate twice that of workers who performed 50-150 lifts per day. However, workers who performed greater than 150 lifts per day exhibited the highest incidence rate, roughly two-thirds higher than the low frequency group. Marras et al (1993) found lift rates to be significantly different between the high risk and low risk groups. The average lifting frequency for the high-risk group was 2.9 lifts/minute while the average for the low risk group was 2.0 lifts/minute. This large relative difference, however, did not yield an odds ratio significantly different from 1.0, most likely caused by the large range of lifting frequencies experienced by the members of both groups. The study performed by Kelsey et al (1984) included lifting frequency in their analysis of acute prolapsed intervertebral discs. They found that people who lifted loads of at least 25 lb. at a frequency greater than 25 times/day were roughly 3 1/2 times more likely to suffer a prolapsed intervertebral disc than people who performed no lifting in their jobs. However, this effect can be attributed to the weight of the object (25 lb. vs. no load), the frequency (25 times/day vs. 0 times/day) of lifting, or a combination of the two factors. Their analysis could not distinguish between the effects of load and frequency in this study.

The risk factors discussed earlier, including load weight and posture, have been simulated on cadaver spinal segments (e.g., Adams and Hutton, 1982; Adams and Hutton, 1983) and have produced repeatable results. However, the temporal effects of the cyclic spinal loading and the repeated flexion/extension cycles that occur with frequent lifting are much harder to simulate in cadaveric specimens, because of the physiological differences between cadaveric and live spinal segments (Keller et al, 1990). The effects of cyclic loading, including repeated microtrauma to the tissues, enhanced fluid flow, and enhanced blood flow, cannot be simulated *in vitro*. Therefore, there is a lack of understanding as to how the spine reacts biomechanically to different loading frequencies and durations. In fact, the National Institute for Occupational Safety and Health (NIOSH) does not directly account for the biomechanical effects of different lifting frequencies in its equation for the Recommended Weight of Lift (RWL) for lifting tasks. NIOSH only includes lifting frequency as it relates to minimizing energy expenditure and localized muscle fatigue during manual material handling tasks (Waters et al, 1993).

Biomechanists who research the effects of temporal factors on low back injury often do so indirectly by looking at what leads to localized muscle fatigue. More recent research has begun to consider how localized muscle fatigue of the trunk extensor musculature may affect the biomechanical risk factors for low back injury. The following sections will provide a more detailed discussion of the literature related to localized muscle fatigue, proprioception and the effects of localized muscle fatigue on proprioception.

Localized Muscle Fatigue

Localized muscle fatigue can be used as an indicator of potential musculoskeletal injury risk since higher force, movement speed, and lifting frequency are associated with both a higher risk of injury and a decreased time to fatigue. However, it is uncertain if the direct effects of localized muscle fatigue are damaging to muscles and joints. One finding has shown that muscle fatigue may make muscles more susceptible to muscular strain injuries by decreasing the amount of energy the muscle can absorb before its failure (Mair et al, 1996). Also, it has been shown that extensive eccentric exercise leads to microscopic structural damage to the muscles (Clarkson and Newham, 1995). However, there is very little research on the direct impact of muscle fatigue on the risk of musculoskeletal injuries.

In contrast, muscle fatigue has been shown to have many secondary effects of concern to industrial ergonomists. These include altered lifting patterns with quadriceps fatigue (Trafimow et al, 1988), decreased movement precision (Lance and Chaffin, 1971; Parnianpour et al, 1988), increased co-contraction/antagonistic muscle activity (Psek and Cafarelli, 1993; Potvin and O'Brien, 1998), and altered spinal loading patterns during lifting tasks (Sparto et al, 1997b, 1997b; Dolan and Adams, 1998). Therefore, it logically follows that localized muscle fatigue can be utilized as a partial litmus test for the existence of a poor match between a worker's capacity and a job's demands.

EMG and Fatigue

Surface electromyography (EMG) has been used extensively in previous research on localized muscle fatigue, because of its ability to non-invasively determine the state of fatigue in a muscle. Integrated EMG has been shown to be proportional to muscular force output in isometric (Lippold, 1952) and isokinetic (Bigland and Lippold, 1956) exertions while parameters which can be derived from the raw EMG signal (median frequency, mean power frequency, and zero-crossing rate) have been found to be reliable in determining the state of fatigue of a muscle (Basmajian and DeLuca, 1985). The physiological explanation for the use of the raw EMG in the determination of muscle fatigue is based in the two general types of fibers found in muscle, fast twitch fibers and slow twitch fibers. Fast twitch fibers activate at a higher frequency than slow twitch muscles, but they also fatigue faster. Therefore, when a muscle starts to fatigue, the fast twitch fibers start fatiguing first, then the slow twitch fibers increase their activation levels within the muscle. Another effect of fatigue is the slowing of the rate of twitch generation and relaxation, reducing the number of pulses per time period possible. Thirdly, the conduction velocity also changes with fatigue. The net result of these effects is that the distribution of firing frequencies of all the motor units shifts towards the lower frequencies. This change in firing frequency can be calculated using the raw EMG signal which is then decomposed into its constituent firing frequencies using the Fast Fourier Transform (FFT) algorithm. The result is a decrease in the mean or median firing frequency, the most commonly cited statistics for the indication of localized muscle fatigue.

Low Back Muscle Fatigue

The effects of muscle fatigue on the low back have been researched and have shown both kinematic and biomechanical loading effects of low back muscle fatigue during repetitive lifting tasks. Whole-body kinematic effects of erector spinae muscle fatigue have shown a transfer of a portion of the external torque about the lumbosacral joint lifting load from the knees to the low back in repeated sagittally-symmetric lifting (Marras and Granata, 1997; Sparto et al, 1997c, Dolan and Adams, 1998). These changes in kinematic parameters alter the reaction forces experienced in the spine after the onset of localized muscle fatigue in any of the involved muscles.

The effect of repetitive lifting on whole-body lifting kinematics has been studied and has focused on the relationship between the amount of knee flexion and trunk flexion at the beginning and the end of the repetitive lifting bout. The general finding of this research is a decrease in knee flexion and increase in trunk flexion as a function of time into a repetitive lifting task (Hagen et al, 1995; Sparto et al, 1997c; Dolan and Adams, 1998; Dieen, et al, 1998). Hagen et al (1995) found that the lifting frequency, not load weight, increased trunk motion and decreased knee motion during repeated sagittally-symmetric lifts using the squat technique. This change did not occur when the subjects started from the beginning of the task using a stoop technique. Sparto et al (1997c) found a 9^{o} decrease in peak knee flexion (measured as the included angle between the shank and the thigh) from 124^{o} to 135^{o} along with smaller increases in peak hip flexion and peak lumbar flexion after a short (130+/-37 seconds), but very fast (37 +/- 4 lifts/minute lift rate), lifting task. Dieen et al (1998) found an increase in peak trunk flexion angle and a decrease in knee flexion at the point of lift after a trial of repetitive sagittally-symmetric barbell lifting. Finally, Sparto et al (1998) found that peak knee flexion angle decreased 11% while peak trunk flexion angle increased 7.4% after a bout of maximal dynamic lifting exertions. Marras and Granata (1997) found that after a 5-hour depalletization task (lifting boxes off a loaded warehouse pallet), the peak trunk flexion angle and peak trunk angular velocity both decreased as a function of time during "standard test lifts" while the peak sagittal angle and velocity of the hip increased. These standard test lifts were sagittally-symmetric to control for box location at the origin and destination and were not part of the depalletization task. The various methodological differences between the Marras and Granata (1997) study and the Sparto et al studies (1997b, 1998), such as speed or lifting and duration of the experiment, may account for some of the contradictory findings.

This alteration in lifting style suggests that the subjects may be subconsciously trying to maximize the time until total exhaustion. Three possible theories may explain this phenomenon. First, this could be an effort to minimize energy expenditure as Garg and Saxena (1979) have shown that "squat lifting" (i.e., using more knee motion and less trunk motion) requires much more energy than "stoop lifting" (i.e., using less knee motion and more trunk motion). Secondly, this alteration can be an effort to reduce the load on the knee extensors. Trafimow et al (1993) showed that a quadriceps muscle fatiguing task alters the lifting kinematics during sagittally-symmetric lifting. They found peak knee flexion angle during lifting decreased after the fatiguing task. Thirdly, the subject may be trying to reduce the load on the lumbosacral spine. As Dieen et al (1996) have shown, "squat lifting" yields higher external torques about the lumbosacral joint than does "stoop lifting." Any of these explanations is plausible and they are not necessarily mutually exclusive.

An alternative analysis that researchers have used in the kinematic analysis of repetitive lifting is phase analysis. This type of analysis looks at the timing or coordination between motion of two different joints, usually the trunk (or hip) and the knee. Dieen et al (1996) found that even after short bouts of repetitive sagittally-symmetric lifting (5 minutes at 15 lifts/minute), the timing of knee extension relative to hip extension was altered relative to the beginning of the task. This decrease in phase lag between the knee and hip extension was attributed to the fatigue of the knee extensor muscles during the repetitive squat-lifting bout. The same was not found for repetitive stoop-lifting bouts. Later, Dieen et al (1998) found that a larger group of subjects exhibited a decrease in phase lag between hip extension and trunk extension. Sparto et al (1997c) found that the timing of both hip and lumbar spine extension skew towards the beginning of the lift and the phase lag between these two measures also increased, meaning the timing of lumbar spine extension fell further behind hip extension as fatigue emerged. While all of this research is very important, the extent to which these findings apply to asymmetric lifting is unknown. As previously discussed, asymmetric lifting has been shown to be related to higher risk of injury (Marras et al, 1993; Fathallah et al, 1998b).

In addition to these kinematic changes as a function of fatigue, internal forces within the spine have also been found to change with the onset of localized muscle fatigue during sagittally-symmetric lifting (Sparto et al, 1997b; Dolan and Adams, 1998). Marras and Granata (1997) showed that repetitive lifting over a 5-hour period yielded changes in recruitment patterns of the trunk, such that a decrease in spine compression occurred along with an increase in antero-posterior shear during sagittally-symmetric lifting. Muscle fatigue was not monitored in this study, but it can be hypothesized that fatigue may have started to occur since these results agree with similar research. Sparto et al (1997b) have shown with a transverse cutting plane EMG-based biomechanical model of the trunk that the loads on the spine changed after repeated isometric fatiguing trunk extension exertions. Sparto and Parnianpour (1998) extended this research by applying the EMG-based model to repetitive sagittally-symmetric isokinetic trunk extensions, but found no significant changes in intervertebral joint reaction forces coinciding with significant changes in substitution patterns. It seems that higher muscle substitution effects on the joint reaction forces would be expected from the dynamic tasks (Marras, 1998). However, the choice of a sagittally-symmetric task may partially explain the lack of an effect, and the importance of considering asymmetric postures during manual lifting tasks will be discussed next.

Asymmetric tasks that involve the oblique muscles more prominently may show more dramatic changes in muscle substitution patterns and co-contraction. Thelen et al (1995) showed that co-contraction ratios are higher for twisting and lateral bending exertions than for extension exertions. Also, Fathallah et al (1998a) showed the importance of complex motions on combined shear and compression loading of the spine. Extension of the spine with simultaneous twisting and/or lateral bending provides the lumbar trunk musculature with many more degrees of freedom than the case of simple sagittally-symmetric trunk extension. As Lavender et al (1992) have shown qualitatively and Potvin and O'Brien (1998) have shown quantitatively, the activation levels of trunk muscles are highly correlated with their moment arms in the plane of the intended action. This research shows the importance of considering the biomechanics of asymmetric trunk motion during manual materials handling research.

Sparto et al (1997b) found that the primary shear-generating muscles (external oblique and latissimus dorsi muscles) increased their activity after fatiguing sagittally-symmetric isometric exertions. These shear-generating muscles are even more active in asymmetric trunk extensions. Potvin and O'Brien (1998) showed that spinal loading increases with fatiguing lateral bend exertions (Potvin and O'Brien, 1998) and fatiguing axial rotation exertions (O'Brien and Potvin, 1997). This research suggests that the more important spinal loading condition of non-sagittally symmetric tasks should be analyzed for changes in loading due to fatiguing lifting tasks.

Proprioception

Proprioception can be broken into the sensation of position, the sensation of movement, and the sensation of force or effort level. Each of these sensations is vital to controlled motor performance. To perform a goal-oriented motion, a person must understand his/her present body position, know the relative exertion levels of the active muscles in order to integrate them properly, then guide his/her body parts through space, hone in on the target, and stop at the target. Proprioception plays an important role throughout this activity. First, the subject must obtain knowledge about the positions of the joints that will be involved in the activity. Then, the muscles that will move the joint(s) must activate in such a pattern (speed, level, and timing) so as to induce the desired motion. Next, the person must assess the progress of the activity and make any necessary adjustments to the muscle activation parameters. Finally, the person must initiate the deceleration phase of the activity when the specific activity ceases by slowing the motion and eventually stopping (or redirecting the motion).

Muscles primarily contain three different proprioceptive mechanisms to sense the physical state of the body parts they affect. Muscle spindles imbedded within the muscles, Golgi tendon organs in the tendons, and free nerve endings throughout the muscle all provide information about the body parts they affect.

Muscle spindles are sensitive to stretch in the muscle in which they are imbedded, thereby providing information which can be used to ascertain the position of the relevant body part. Their sensitivity can be altered via contraction of the intrafusal muscle fibers, which are housed within the muscle spindle. The activation of the spindles often affects the surrounding muscle via increased agonist muscle activity and decreased antagonist muscle activity via an interneuron (McComas, 1996). As an example, the performance of the muscle spindles is what is tested during a common patellar reflex test performed by a physician. When the tendon is hit with the instrument, the muscle lengthens. This lengthening is sensed by the muscle spindles, fed back to the quadriceps muscle group via the Group Ia afferent nerves, and then a reflex contraction of the quadriceps occurs (McComas, 1996). The reflex causes the muscle to contract in order to restore the muscle to its original length.

Golgi tendon organs are important in the control of motion because they sense the amount of tension in the muscle, due to either passive stretching or active contraction of the muscle. Unlike the muscle spindles, the Golgi tendon organs are located in the tendons of muscles and are attached to extrafusal muscle fibers (McComas, 1996). The nerve "twigs" of the Golgi tendon organ are interwoven with the collagen matrix in the tendon and when the muscle exerts tension on the tendon, the collagen compresses the nerve twigs and the Group 1B afferent nerve to which the nerve twigs are connected is discharged (Swett and Schoutz, 1975). These receptors act in a protective manner, inhibiting activation of the motoneurons of the same muscle while exciting those in the antagonist muscles.

The last category of receptors present in muscles is the free nerve endings. They vary in their function, but they can sense mechanical, chemical, or nociceptive stimuli. McComas (1996) notes that the free nerve endings can sense changes in ion concentration such as K+ or lactic acid in the local environment. This can be important in a fatiguing situation as the pH changes with the onset of fatigue due to the buildup of lactic acid. Also, free nerve endings can sense mechanical stimuli from muscle contraction and stretching. Finally, free nerve endings can sense nociceptive stimuli due to tissue damage. The research of the free nerve endings has been limited in proprioception and more applied to pain sensation. The one relevant article found was by McNair and Heine (1999) who found that trunk repositioning accuracy and precision improved when subjects wore a neoprene lumbar support (average absolute error (AAE) = 3.03^{o} and standard deviation = 3.81^{o}) as compared to the no-brace condition (AAE = 3.62^{o} and standard deviation = 4.41^{o}). Since all the subjects were blindfolded during the testing, it can be argued that the free nerve endings which sense external pressure on the skin were the primary proprioceptive mechanism responsible for this improvement.

Proprioception in Ergonomics

Proprioception is important in the area of industrial ergonomics. In a manual working environment, a single task may be performed thousands of times a day. For example, a manual materials handler who is lifting packages at a rate of 8 per minute will perform about 3360 lifts in a normal 8 hour shift (assuming an hour for lunch and breaks). Each of these lifts requires targeted movements and precise placement of the materials being handled. It is apparent that the proprioceptive mechanisms in this person's nervous system are continuously working to allow the worker to perform his/her job tasks accurately, easily, and quickly. These task requirements will be discussed in terms of proprioceptive needs.

Accuracy is a key ingredient in any job as product quality is inherently linked to worker accuracy. The accuracy of a carpenter when nailing plywood flooring to a floor joist will affect the quality of the completed floor if some nails miss the joist. The accuracy of a person doing word processing is a key productivity issue since correcting mistakes is such a time-consuming activity. This accuracy is dependent upon precise knowledge about the location of the hands at any point in time. Fitts (1954) found a highly repeatable linear relationship between the speed of an activity and the log of the activity's required precision (usually measured in target size). This speed-accuracy tradeoff is of interest to industrial engineers who must design workplaces to make sure that the workers can perform the highest level of quality in the shortest time.

Workers are constantly learning new tasks. As assembly lines change to adapt to new product lines and as engineering initiatives such as job rotation and flexible manufacturing are implemented, workers must learn new tasks. Proprioception plays an important role in this learning process. As a new task is learned, the worker must be able to sense his/her body position while learning the task. S/he must be able to reduce the level of activity in the involved muscles after overshoot and increase activity after undershoot. For workers performing manual materials handling tasks, the task learning can be related to larger issues such as how much trunk flexion vs. knee flexion to utilize or what speed to work at so as to minimize fatigue.

As workers learn new job tasks, improving movement efficiency becomes important, especially for jobs requiring high performance. The elimination of wasted actions, knowing when to ask for assistance, and optimized motion patterns all can improve a worker's efficiency. However, in order to become more efficient, the inefficiencies must be recognized. For example, if a manual materials handler repeatedly goes home at the end of a day with low back discomfort, s/he will be more motivated to search for more efficient work patterns. This level of discomfort, however, must be sensed and monitored closely to see if newer techniques decrease the level of discomfort. Improvement at this point can be very minute, so the level of precision with which the physical effort must be sensed is quite high.

Proprioception of the Spine

Although proprioception is important in all body parts for workers, the issues related to the lumbar spine and the implications for manual materials handling tasks will be the focus of this section and this research. The spine has become a popular anatomical location to test proprioception. Physical therapists have tried to determine the effectiveness of rehabilitation treatment protocols on the neuromuscular system using proprioceptive testing by understanding the level of neuromuscular control regained by the patient (Maffey-Ward et al, 1996).

The trunk is a very interesting application of proprioception due to its complex musculoskeletal anatomy. The many joints, degrees of freedom, ligaments, and muscles that are involved in moving the trunk provide a highly complex subsystem to monitor. The intervertebral discs are not very well innervated (Bogduk, 1983) and the lumbar and thoracic intervertebral discs contain low numbers of mechanoreceptors (McLain and Pickar, 1998). Therefore, the body must rely heavily on the spinal musculature to provide proprioceptive information to the neuromuscular system. The diverse angles of pull and moment arms of the trunk muscles provide for a considerable amount of potential information from the muscle spindles and Golgi tendon organs. Several studies have measured proprioception in the different planes of motion using trunk repositioning accuracy, which is defined as the difference between a target trunk angle in a given plane and the actual trunk angle in subsequent trials. Swinkels and Dolan (2000) found better trunk repositioning accuracy in the lateral plane than in the sagittal plane. Newcomer et al (2000) found that the worst repositioning precision occurred in the sagittal plane. McGlashen et al (1991) found repositioning accuracy was better (although not statistically significant) in the lateral plane than in the sagittal plane. The repositioning precision and accuracy are most likely better in the lateral plane than in the sagittal plane due to the difference in the moment arms of the muscles experiencing the greatest stretch. During bending activities, the primary sense of position change is coming from the outermost muscles, the oblique abdominal muscles in lateral bending and the erector spinae in forward flexion. The distance between the center of the oblique muscles and the intervertebral joints is greater than the distance between the erector spinae muscles and the intervertebral joints. Thus, the change in length of the oblique muscles is larger for a given change in the intervertebral joint angle in the lateral plane than the amount of stretch in the erector spinae for the same change in the intervertebral joint angle in the sagittal plane.

Previous researchers looking at trunk proprioception have used several different techniques to assess trunk proprioception. Position sense has been tested by either active posture replication or passive posture detection. Active posture replication tests require the subjects to simply replicate an initial trunk posture in one of the three planes of motion while a motion analysis system measures the trunk angle (Taylor and McCloskey, 1990; Parkhurst and Burnett, 1994; Swinkels and Dolan, 1998; Gill and Callahan, 1998; McNair and Heine, 1999; Newcomer et al, 2000). Passive posture detection utilizes some type of motorized device that moves the subject's trunk slowly at a constant angular velocity and the subject hits a trigger when s/he feels that s/he has hit the target posture (Taylor and McCloskey, 1990; McGlashen et al, 1991; Parkhurst and Burnett, 1994). Both repositioning accuracy (difference between the mean of replication postures and the target posture) and repositioning precision (standard deviation of the replication postures) have been evaluated in this area of research. Other measurements such as minimal angular displacement required to sense a change in position (Taylor and McCloskey, 1990; McGlashen et al, 1991; Parkhurst and Burnett, 1994; Taimela et al, 1999) and direction of movement (Parkhurst and Burnett, 1994) have also been used in proprioception research. Table 2 summarizes the previous research that has been performed on the proprioception of the trunk.

Table 2: Summary of trunk repositioning research.

Reference	Subjects	Test Posture	Trunk Angles Tested	Movement Speed	Reps / Set	Measurement Device	Average Absolute Error (AAE)
Taylor and McCloskey, 1990	8 healthy students	Sitting	5^o and 15^o	Subject choice	40	Potentiometer device on shoulders, measured relative to chair	Twist: 2.85^o
McGlashen et al, 1991	10 healthy men	Standing	~20^o	Subject choice	3	Linear displacement markers at L3, T11, T6, T1, and Head	Flexion (T1): 1.2^o Lat. (T1): 0.8^o
Parkhurst and Burnett, 1994	88 EMS workers and firefighters (33 healthy, 55 previous injury)	In apparatus	5^o	Subject choice	1	Linear displacement markers on apparatus	Flexion: 23.3 Lateral: 20.4 Twist: 16.5 ***
Swinkels and Dolan, 1998	20 healthy adults	Standing	½ max ROM	Not reported	1	Fastrak sensors on T1, T7, L1, and S2	Flexion (T1): 3.9^o Lat. (T1): 3.1^o
Gill and Callaghan, 1998	20 healthy adults and 20 LBP patients	Standing & kneeling	20^o (standing)	3 seconds / rep	10	LMM	Flexion: 4.4^o
McNair and Heine, 1999	40 healthy adults, braced and unbraced	Standing	6 random flexion angles		1	LMM	Flexion (unbraced): 3.6^o
Newcomer et al, 2000	20 healthy adults and 20 LBP patients	Standing	50% max ROM	5 Seconds to obtain position	3	3Space markers on L1 and S1	Flexion: 2.6^o Lateral: 1.4^o Twist: 1.0^o
Swinkels and Dolan, 2000	20 healthy adults	Standing	1/3, 1/2, and 2/3 max ROM	6 seconds / rep	3	Fastrak markers on T1, T7, L1 and S2	Flexion (L1): 3.3^o, 4.3^o, 4.2^o Lat. (L1): 1.6^o, 1.6^o, 1.7^o
Iwasa et al, 2005	25 healthy adult males	Standing	1/3, 1/2, and 2/3 max ROM	Not reported	3	Fastrak sensors on T9 and S2	Flexion: 2.5^o /3.4^o Lateral: 1.6^o /2.1^o (unfatigued/fatigued)

*** = Measures were average error, not average absolute error; measurements in mm, no angle measurements given

NOTE: All previous research was performed with subjects' eyes closed or blindfolded with one exception. Newcomer et al (2000) tested subjects with eyes open and with eyes closed.

In terms of force sensation, some research has been done in the area of biofeedback and pain regulation (Peck and Kraft, 1977; Flor et al (1992), where the sensation and regulation of muscle activity can reduce symptoms with tension headaches (Peck and Kraft, 1977). Flor et al (1992) found that LBP patients were not as accurate in estimating their level of erector spinae muscle activity as normal subjects. Descarreaux et al (2004) used a force replication procedure to evaluate the motor control of low back patients and healthy subjects. They found that subjects with low back pain could produce isometric forces with similar accuracy and precision as healthy subjects, but the rise time was slower in patients.

Several different measurement techniques have been used to measure the different proprioceptive elements of the spine. Repositioning accuracy and precision have been measured in several different ways. The Lumbar Motion Monitor (LMM) has been used by Gill and Callahan (1998). The 3Space Fastrak electromagnetic marker tracking systems has also been used in previous research (Newcomer et al, 2000; Swinkels and Dolan, 2000). The Electromagnetic tracking systems are very accurate (0.3" RMS accuracy and 0.03" resolution; Ascension Technology Corporation web page, 2000), but they tend to have interference problems with metal objects that are within the magnetic field. Other specialized devices have been fabricated to measure trunk repositioning such as the setup used by McGlashen et al (1991) which measured the position at various vertebral levels.

Several researchers have compared LBP patients to healthy individuals. Parkhurst and Burnett (1994) found no difference in motion threshold detection, motion direction perception, or repositioning accuracy of the trunk using a passive motion apparatus. However, their injured subjects were firemen who were currently working, but who had an

LBI previously. Gill and Callaghan (1998) found better repositioning accuracy (in terms of average absolute error) for normals (4.5^{o}) than for LBP patients (6.7^{o}). They used the LMM to measure the repeatability of 10 continuous trunk flexions from standing to 20^{o} in a period of 30 seconds. Newcomer et al (2000) found no difference in proprioception in terms of repositioning accuracy between LBP patients and normals.

Flor et al (1992) found that LBP patients were worse than normals at gauging static muscle activity for both the masseter muscle (unaffected muscle) and the erector spinae muscle (effected muscle). Patients typically exerted EMG values less than the target value. Therefore, there is no consensus as to the effect of LBP on trunk proprioception, if any. The differences in subject groups or proprioception measurement technique in these studies may partly explain these differences.

Proprioception and Fatigue

Localized muscle fatigue has been found to affect proprioception in various ways. The sense of position, movement, and force all are altered after the onset of fatigue. These effects have been shown in various joint systems.

The sensation of force has been shown to increase with localized muscle fatigue (Jones, 1986). Jones concluded that sensations of muscle force are due more to efferent activity than to peripheral proprioceptive feedback. This conclusion was based on the fact that the body seems better able to sense the relative effort put forth in an exertion (e.g., in terms of % MVC) than sense the absolute force exhibited. She mentioned that findings have shown better weight discrimination with active lifting than with just passive force application. She cites several references that show that the perceived force increases with time in a constant-force exertion. Jones and Hunter (1983a, 1983b) used a matching force experiment by fatiguing one arm's elbow flexors until the maximal force output was less than 50% of the original maximum force and trying to periodically match the force level in the other arm's elbow flexors. The fresh arm's elbow flexion force increased linearly with time by the same rate across 35%, 50%, and 65% MVC on different days. This correlated well with the fatigued biceps' full-wave rectified EMG value ($r = 0.95$; Jones and Hunter, 1983a). This is an example where the body was apparently sensing the relative effort rather than the actual force being lifted. Therefore, it seems that the body's ability to sense an increase in effort level could be used as an indicator of the beginning of muscle fatigue as determined by an increase in muscle activity to compensate for a decrease in maximal muscle stress per cross-sectional area (N/cm^2).

Several studies have looked at the proprioceptive performance of the shoulder joint and the knee joint before and after fatigue. Sterner et al (1998) looked at repositioning precision (reproduction of active and passive positioning) and passive motion threshold detection in internal and external rotation at the shoulder before and after fatigue. They had subjects perform continuous maximal concentric/concentric internal/external shoulder rotations until the maximal force fell below 50% of the original force. They used a 2 minute window after the end of the test from which to test the fatigued condition. They found no effect of fatigue in any of the four measurements of interest. They concluded that the quick recovery from the exercise might have caused this discrepancy. However, Carpenter et al (1998) found results opposite those of Sterner et al (1998). They found that motion threshold detection increased (i.e., worsened) by 73% from 0.92^{o} to 1.59^{o} when rotated at 1^{o}/sec after repeated maximal internal/external shoulder rotation. The difference between these two studies could lie in the fact that the former experiment performed four tests after fatigue and the latter experiment just tested motion detection before and after fatigue, thus minimizing the recovery period experienced by the subjects. Pedersen et al (1999) was the only experiment that tested the sense of motion. They used a forced-choice paradigm to have subjects compare a test velocity (randomly chosen from a normal distribution between 40^o/sec and 60^o/sec) to a reference velocity (50^o/sec). The activity was passive (torque < 5Nm) horizontal extension of the shoulder. They compared the effect of performing fatiguing maximal horizontal flexion/extensions and submaximal (10% MVC) exertions. They found that the probability of correctly choosing the faster velocity was lower during and after the fatiguing maximal exertions. This showed that ones' judgment of movement angular velocity decreases with fatigue. Skinner et al (1986) found worse passive repositioning precision of the knee joint after fatiguing quadriceps exercise. The repositioning precision worsened by more than a degree, going from 2.90^o to 3.97^o.

The most relevant previous research found in the literature were studies performed by Taimela et al (1999) and Iwasa et al (2005). Taimela et al (1999) tested the effect of trunk extensor muscle fatigue on the trunk rotation motion threshold detection on both normals and LBP patients. They found that both groups exhibited a significant increase (worsening) of the amount of trunk rotation required to detect a change in trunk position. Interestingly, the data showed dramatic improvement immediately after the test. So recovery, as suggested by Sterner et al (1998) above, may play an important part in the effect of fatigue on joint proprioception. More recently, Iwasa et al (2005) performed a study of the effects of erector spinae muscle fatigue on repositioning error in the sagittal and coronal planes. They induced fatigue in their subjects using sagittally-symmetric isokinetic trunk extensions performed at 30 ^{o}/sec at maximal load until they could no longer maintain a load of at least half of the original maximum. The mean endurance time for this task was 308 seconds, so their fatigue was generated using a relatively short duration, high load exercise. Using the same protocol as Swinkels and Dolan (2000), they found an increase in trunk repositioning error in both flexion (from 1.9^{o} to 3.1^{o} at 1/3 maximum forward flexion angle) and lateral bending (from 1.8^{o} to 2.4^{o} at 1/3 maximum lateral bending angle) postures after the bout of fatiguing isokinetic trunk extensions. While these effects were only significant at the 1/3 maximal angles and not at the 1/2 and 2/3 maximal angles, the results in the other angles showed non-significant increases in error.

To date, however, no single study has evaluated the effects of trunk muscle fatigue on active trunk angle positioning accuracy and precision in all three planes of motion. While the previous research suggests that erector spinae muscle fatigue would reduce this active trunk repositioning accuracy and precision in all three planes, no research has shown this to date.

Sudden Unexpected Loading

Sudden loading of the trunk has been identified as a risk factor for low back injuries (Troup et al, 1981, Manning et al, 1984; Omino and Hayashi, 1992; Cholewicki et al, 2002). Researchers have looked at the reaction to sudden loading in terms of muscle activation levels, spinal loading, trunk stiffness, and muscle onset and shutoff times. Lavender et al (1989) found a linear relationship between preview time of a sudden load and both the peak and mean EMG values. Later, Lavender et al (1993) found that although subjects typically increased erector spinae EMG before the loads were applied, the activation of the other trunk muscles was highly variable. Both Marras et al (1987) and Mannion et al (2000) found increased loads on the spine due to loads that were unexpected compared to similar loads that were expected.

More recently, researchers have used the muscle onset times and activation levels along with anatomical data to biomechanically model the stiffness of the trunk as a function of sudden loading (Granata et al, 2004; Lawrence et al, 2005). Granata et al (2004) found that fatigue decreased trunk stiffness during responses to sudden loads. However, Lawrence et al (2005) found no effect of fatigue on peak or mean trunk stiffness in sudden loading responses of the trunk. Since this line of research is rather new there is no a full understanding of the effect of fatigue on trunk stiffness.

Previous research on the muscle onset times in the trunk as a response to sudden loading shows that muscle onset times increase with low back muscle fatigue (Wilder et al, 1996; Magnusson et al, 1997). Also, low back pain patients have shown increased muscle onset times and muscle shutoff times when compared to normal control subjects (Radebold et al, 2000). It is impossible to tell from this study whether the increased onset and shutoff times were factors that contributed to their condition or an adaptive response to the condition. Balestra et al (1992) showed that localized muscle fatigue reduced the EMG amplitude of the short latency and medium latency components of the stretch reflex of the first dorsal interosseus muscle of the hand. However, they found no difference in the onset time of these different components.

Brief Summary of the Literature

This review of the literature related to biomechanics, fatigue, and proprioception has identified several important research voids that the current research seeks to fill. First, while the literature has shown that proprioceptive mechanisms in the musculoskeletal system can be compromised when exposed to a repetitive fatiguing exertion, it is not clear that this response can be seen in levels of exertion intensity and duty cycle found in more realistic work tasks. Second, while it has been shown that during a repetitive lifting bout, there are changes in the whole body kinematics that affect the distribution of loading between the low back and the lower extremities, it is not clear what role the proprioceptive system plays in this response. Third, while it has been shown that during repeated trunk extension exertions the relative activation of the various trunk muscles changes (thereby altering spinal loading), it is not clear what role the proprioceptive system plays in this response. These observations have led to the development of a proposed theoretical model for the inter-relationships between fatigue, proprioception, lifting kinematics and muscle activation patterns.

THEORETICAL MODEL

The theoretical model that was originally proposed is shown in Figure 2. This model was then updated based on feedback from the research proposal and the updated model is shown in Figure 3, which has the addition of trunk force and reaction time added. This model hypothesizes a relationship between localized muscle fatigue of the trunk extensors and changes in the proprioceptive capabilities of the musculoskeletal system of the low back (decreased position sensation and decreased force exertion level sensation). It is further hypothesized that these changes in proprioceptive capabilities can influence biomechanical performance measures. First, decreased force sensation impacts the trunk muscle coactivation patterns through over-shoot or under- shoot of force by the trunk extensors. Second, decreased trunk position sensation can affects the trunk kinematics used to perform a lifting task. Third, the error in the proprioceptive measures increases the variability of the muscle activation patterns and the motion patterns which, in turn, lead to increased peak loads on the upper end of the distribution of the resultant intervertebral joint reaction forces. Fourth, the maximal trunk extension torque is hypothesized to impact the kinetic portions of the model. Lastly, the reaction time is hypothesized to impact both the kinematic and kinetic portions of the model.

In this model, these effects are expected in both the mean (a measure of accuracy) as well as the variability (a measure of precision) of these measures. While the underlying theoretical model suggests causation between muscle fatigue and the subsequent effects on positional and muscle activity accuracy and precision, causation was not involved in the hypothesis testing of this research. The experiments in this research simply looked for associations between the level of muscle fatigue and the subsequent effects on proprioception, lifting kinematics and trunk muscle coactivation. Far more testing would be involved to show any level of causation and is far outside of the scope of this research.

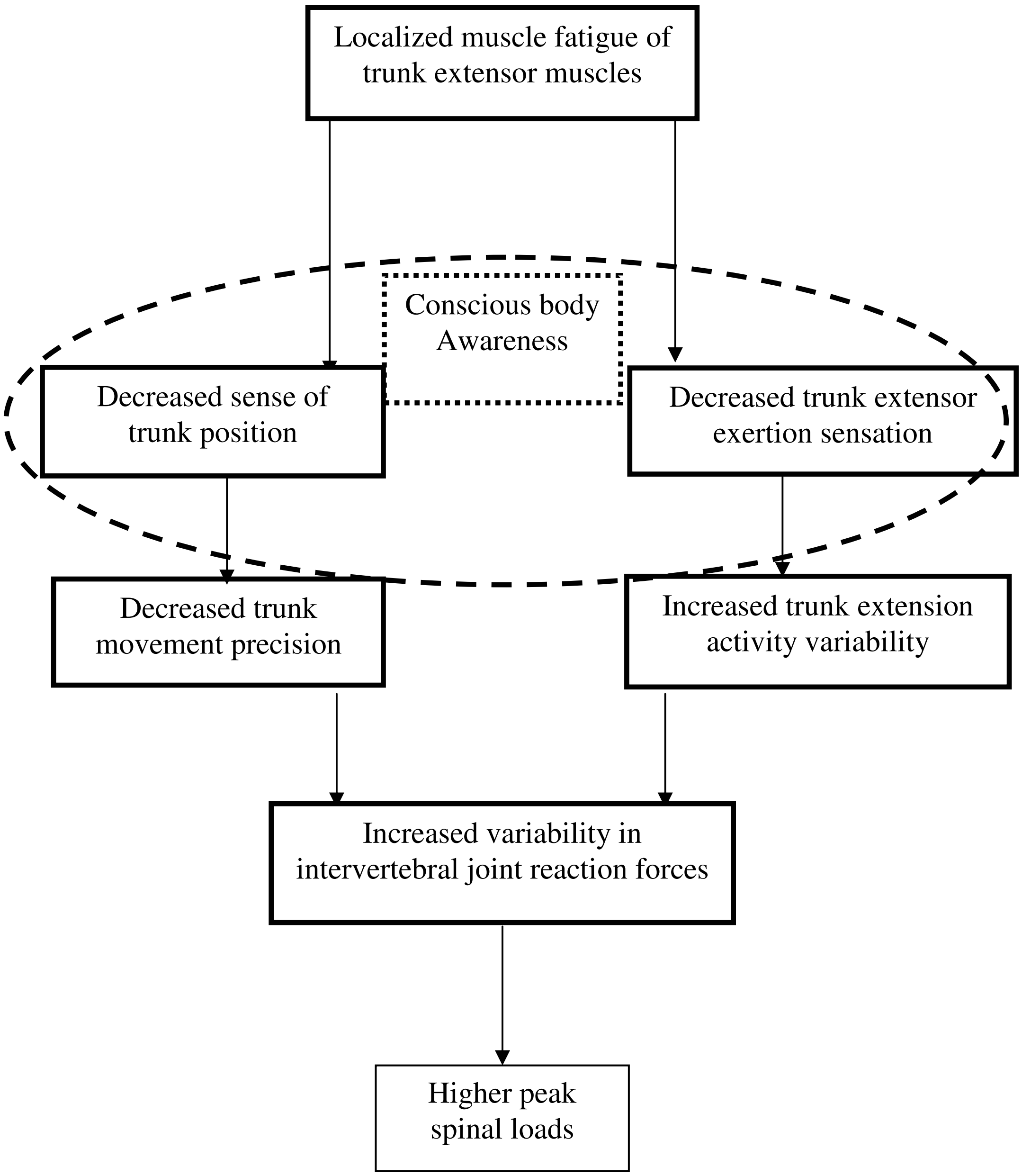

Figure 2: Graphic representation of the original proposed model of the effects of repetitive lifting.

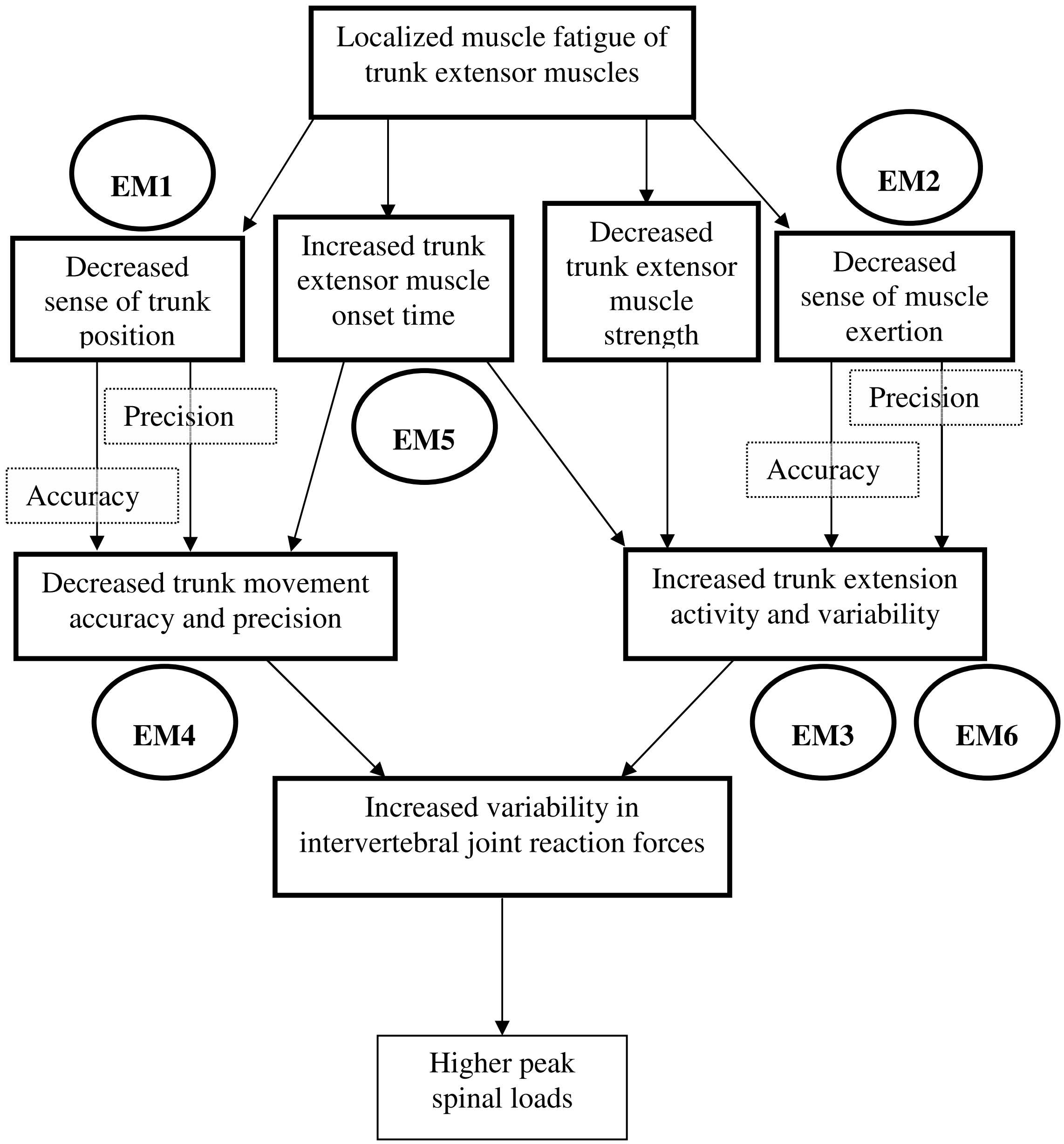

Figure 3: Graphic representation of the proposed model of the effects of repetitive trunk extensions updated after the proposal defense. The numbers in the circles correspond to the experimental modules in the *Methods* section.

RESEARCH GOAL

The overall goal of this research is to investigate the relationship between the sense and control of trunk position with the sense and control of trunk force as a function of muscle fatigue. The specific goal of this research is to evaluate the changes in trunk extensor strength, trunk proprioception, force sensation, trunk muscle activation strategies, muscle onset times, and whole body lifting kinematics as a function of fatigue induced through repetitive trunk extensions. The specific characteristics of these measures include both the mean (which will be referred to as "accuracy" throughout this document) and variability (which will be referred to as "precision" throughout this document) of these measures. The combined effects of the changes in each of these parameters will be discussed in terms of their impact on spinal loading during repetitive lifting activities.

PILOT WORK

Kelaher, D.P., Glasscock, N.F., Mirka, G.A. (2003) The effects of body awareness, movement direction, visual condition, and movement speed on trunk proprioception.

Proceedings of the Human Factors and Ergonomics Society 47th Annual Meeting, p.1299-1302.

THE EFFECTS OF BODY AWARENESS, MOVEMENT DIRECTION, VISUAL CONDITION, AND MOVEMENT SPEED ON TRUNK PROPRIOCEPTION

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Research on the proprioceptive mechanisms of the trunk typically involves trunk repositioning tasks. However, this body of research involves numerous differences in the test methods, results, and interpretations. This research tested the effects of two of the methodological differences, the visual condition and the speed of movement, along with the plane of motion, on the accuracy and precision of trunk repositioning ability in the three primary planes of trunk motion. The results showed significant main effects of plane of motion and visual condition on both repositioning accuracy and precision. Significant interactions between speed and plane of motion and between the visual condition and plane of motion were found for trunk repositioning accuracy. Finally, there was a strong correlation between the overall repositioning precision and the combined score of the Kinesthetic and Proprioceptive Assessment Questionnaire and the Body Awareness Questionnaire.

INTRODUCTION

A key component in any goal-oriented activity is the ability to sense one's position in space. This proprioceptive capability of various joints has received research attention in the rehabilitation literature (Maffey-Ward et al, 1996). However, proprioception is also important in being able to reliably replicate a given occupational task. Trunk repositioning accuracy and precision tests have been used in this research area as a measurement of proprioception of the trunk (Swinkels and Dolan, 1998; Gill and Callahan, 1999). Trunk proprioceptive testing has been used to test the effectiveness of rehabilitation treatment protocols on the neuromuscular system using by understanding the level of neuromuscular control regained by a patient (Maffey-Ward et al, 1996). This research has used a variety of postures (standing, seated, kneeling, prone), movement directions (flexion/extension, lateral bending, axial rotation), measurement systems (electrogoniometer, electromagnetic tracking systems, miscellaneous mechanical devices), number of repetitions (1, 3, 10), movement speeds (slow, fast) and visual conditions (eyes open or eyes closed/ blindfolded). As one might expect, this has yielded a variety of results.

The goal of this study was to analyze the effects of three of these conditions, the visual condition, the plane of motion, and the speed of movement, on trunk repositioning accuracy and precision in all three planes of motion. The secondary goal was to assess the relationship between trunk proprioception performance and body awareness.

METHODS

Six male subjects were recruited for this study. None had any history of low back pain that restricted work or leisure activities. Subjects were asked to perform 144 repetitions of trunk postures in the three primary planes. Independent variables were the repetition speed, the plane of motion, and the visual condition. The fast speed was performed at 3 seconds / rotation while the slow speed was performed at 6 seconds / rotation. The fast speed was chosen so that it did not give the subjects time to fine tune their position while the slow speed did provide the subjects sufficient time to fine tune their position for each repetition. The actual speed of the motion was not controlled or measured. The two visual conditions were simply eyes open or eyes closed. The planes of motion were the three orthogonal planes of trunk motion, trunk flexion, trunk lateral bending, and axial rotation.

The 144 repetitions were broken down into 48 sets of 3 repetitions that were all randomized. This was done to minimize any time-dependent factors such as fatigue that would occur throughout the testing period. At the beginning of the experiment, each subject established his maximal range of motion in each plane of motion. For all of the test postures, subjects were instructed to replicate a posture exactly half of the initial maximal range of motion exertion.

The Motion Star electromagnetic sensor tracking system (Ascension Technologies, Burlington, VT) was used to measure the angle of the trunk during the test motions. Each sensor is capable of measuring translations and rotations in all 3 orthogonal planes of motion. Sensors were taped onto the skin of the back over the C7 spinous process and the S1 level of the sacral crest.

The dependent variables that were analyzed were the average absolute error (AAE) and the variance of the angle of rotation of the C7 bird relative to the upright standing position in each of three planes of motion. The AAE was analyzed using ANOVA and the variance was analyzed using Bartlett's test of homogeneity of variance.
Each subject also took the Body Awareness Questionnaire (BAQ) and the Kinesthetic and Proprioceptive Awareness Questionnaire (KPAQ). The BAQ is an established survey of 18

questions about awareness of various physiological changes. Its validity and reliability has been tested in previous research (Shields et al, 1989). The KPAQ is a recently-developed survey (Glasscock, 2003) of 12 questions regarding the perceived ability to sense biomechanical changes in one's body.

RESULTS

The data showed a significant effect of plane of motion on the accuracy and precision of trunk repositioning. The accuracy in the sagittal plane was significantly worse than in the other two planes while the precision in the axial rotation was worse than in the other two planes. Table 1A shows the accuracy (AAE) and precision (variance) of the trunk repositioning postures.

Table 1A: Accuracy (AAE) and precision (variance) of trunk repositioning (degrees) as a function of plane of motion. Cells with same letter are not significantly different from each other.

	Flexion	Lateral Bend	Axial Rotation
Variance	66.1 (A)	53.9 (A)	110.7 (B)
AAE	12.52 (C)	6.32 (D)	6.04 (D)

There was also a significant effect of visual condition on both the AAE and the variance. Both the AAE and variance were higher with the eyes open than with the eyes closed, meaning the trunk repositioning accuracy and precision were both better with the eyes closed. Further ANOVA tests were performed on variance data by plane of motion since the Bartlett's test can only be used in a one-way ANOVA test. This further analysis showed that only in the twisting postures were variances of the trials significantly different from each other. Table 2A shows the variance data as a function of visual condition separated by plane of motion.

There was also a significant interaction between plane of motion and the speed of motion on the AAE. Separate ANOVA tests showed that the faster speed yielded better accuracy for lateral bending and twisting, but worse accuracy for trunk flexion. Similarly, there was a significant interactive effect between plane of motion and visual condition on the accuracy of the trunk repositioning. The AAE was significantly lower for the eyes closed condition in the lateral bending and the twisting postures, but not in the flexion postures.

Table 2A: Precision (variance) of trunk repositioning (degrees) as a function of visual condition separated by plane of motion. Cells with same letter are not significantly different from each other.

	Flexion	Lateral Bend	Axial Rotation	All Planes
Eyes Open	72.4 (A)	56.4 (B)	135.2 (C)	196.5 (E)
Eyes Closed	60.1 (A)	51.3 (B)	80.6 (D)	156.8 (F)

When the pooled accuracy and precision data was collapsed across conditions and compared to the total survey data, a strong correlation ($r^2 = 0.59$) between total survey score and the variance of trunk position was evident. This relationship, shown in Figure 1A, shows a negative relationship, meaning as the awareness of body position increased (via the survey score) the precision of trunk posture improved (i.e., decreased).

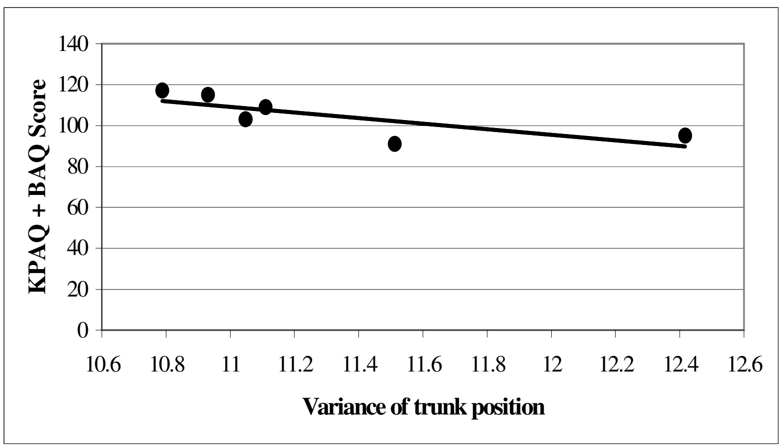

Figure 1A: Relationship between trunk repositioning precision and total score on the KPAQ and BAQ surveys.

DISCUSSION

The finding of worse repositioning accuracy in the sagittal plane supports previous research (McGlashen et al, 1991; Newcomer et al, 2000; Swinkels and Dolan, 2000) as does the non-significant difference between the accuracy in the lateral bending and twisting. These results suggest that there may be a fairly constant subject-dependent ratio between the repositioning precision and the rate of change in length of the muscle, taking into account the angle of pull of the muscle.

The effect of visual condition on the repositioning accuracy and precision is interesting. The only previous research study to compare visual conditions showed no effect of visual condition on the repositioning accuracy (Newcomer et al, 2000) contrary to the findings in this study. However, the study by Newcomer et al provided no detailed data as a function of visual condition. This could be due to the methodological differences in the two studies. This study placed the upper sensor over the C7 spinous process while the other study placed the upper sensor over the L1 spinous process. It is possible that the visual condition could affect the cervical region of the trunk more than the lumbar section.

The relationship between the subjective awareness as measured by the surveys and the overall repositioning precision provides some evidence that people may understand their level of proprioceptive capability. However, a much larger sample size would be needed to validate these results further.

Finally, the findings of this research can be used in decisions on further research on trunk proprioception. Studies which place the upper sensors towards the cervical region may want to perform testing with eyes closed so as to minimize the possibility of subjects looking towards a target and biasing the position data of the cervical sensor.

REFERENCES

Gill, K.P. and Callaghan, M.J. (1998) The measurement of lumbar proprioception in individuals with and without low back pain. *Spine*, 23(3):371-377.

Glasscock, N.F. (2003) *Exploring the relationships between psychosocial factors, biomechanical workstyle, muscle tension, and musculoskeletal discomfort reporting*. Unpublished Dissertation, North Carolina State University.

Maffey-Ward, L., Jull, G., and Wellington, L. (1996) Toward a clinical test of lumbar spine kinesthesia. *Journal of Orthopedic Sports and Physical Therapy*, 24(6):354-358.

McGlashen, K., Ashton-Miller, J.A., Green, M., and Schultz, A.B. (1991) Trunk positioning accuracy in the frontal and sagittal planes. *Journal of Orthopedic Research*, 9(4):576-583.

Newcomer, K., Laskowski, E.R., Yu, B., Larson, D.R., and An, K.-N. (2000) Repositioning error in low back pain. Comparing trunk repositioning error in subjects with chronic low back pain and control subjects. *Spine*, 25(2):245-250.

Shields, S.A., Mallory, M.E., and Simon, A. (1989) The Body Awareness Questionnaire: reliability and validity. *Journal of Personality Assessment,* 53(4):802-815.

Swinkels, A. and Dolan, P. (1998) Regional assessment of joint position sense in the spine. *Spine*, 23(5):590-597.

Swinkels, A. and Dolan, P. (2000) Spinal position sense is independent of the magnitude of movement. *Spine*, 25(1):98-105.

PRELIMINARY WORK

Kelaher, D.P., Mirka, G.A., Sommerich, C.M. (2005) The effects of body awareness, movement direction, visual condition, and movement speed on trunk proprioception.

Proceedings of the Human Factors and Ergonomics Society 49th Annual Meeting, p.1315-1319.

THE EFFECTS OF FATIGUE FROM REPEATED TRUNK EXTENSIONS ON TRUNK MUSCLE ACTIVITY

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Asymmetric trunk postures and trunk dynamics have been implicated as risk factors for occupational low back disorders. Muscle fatigue has also been cited as an ergonomic issue, though not directly linked to an increased risk of low back injury. This paper investigates the effect of trunk extensor muscle fatigue on muscle activity levels during symmetric and asymmetric trunk extension exertions. The results show that muscle fatigue alters both flexor and extensor muscle activity patterns in asymmetric trunk extensions earlier than in sagittally-symmetric trunk extensions. This quickened fatigue response to asymmetric trunk motions could be a contributor to the increased risk of low back injury seen in occupational tasks that include asymmetric trunk lifting.

INTRODUCTION

Research has shown that asymmetric trunk postures and repetitive lifting are risk factors for developing low back disorders (Chaffin and Park, 1973; Punnett et al, 1991; Marras et al, 1993). However, quantifying the cumulative “dose” of low back stress during lifting can be difficult. Muscle fatigue may be an effective indirect measure of cumulative loading. Although erector spinae muscle fatigue has not been directly linked to low back injury risk, there are several effects of muscle fatigue that may impact the risk of low back injury. They include decreased movement precision (Parnianpour et al, 1988), increased co-contraction/antagonistic muscle activity (Potvin and O’Brien, 1998), and altered spinal loading patterns during lifting tasks (Dolan and Adams, 1998). However, these previous papers did not consider asymmetric trunk extensions, focusing instead on single-plane motions. This paper investigates the effect of trunk extensor muscle fatigue on muscle activity during both symmetric and asymmetric trunk extension exertions.

METHODS

Subjects

Nine healthy adult male subjects (18-25 years old) were tested in this experiment. None of the subjects had a history of chronic low back pain that limited their work or leisure activities. All subjects were physically active, but not elite athletes. In order to minimize variability of the data due to large anthropometric differences between subjects, all subjects were in the interquartile range of stature (167.6cm-182.9cm) and weight (63.5kg-90.7kg). Six of the subjects were experimental subjects who performed fatiguing trunk exertions during the experiment while the remaining three subjects did not. Only the data from the six experimental subjects will be provided and discussed in this paper.

Experiment

This experiment was part of a larger study where both kinematic and electromyographic data were collected to analyze the relationships between trunk extensor muscle fatigue, trunk proprioception, and trunk biomechanics. The entire experiment required three sessions on separate days within approximately one week. The first two sessions were to learn the various tests and the third session was the data collection session. Since all of the data was collected during a single data collection session and the other testing procedures may influence the interpretation of the resulting data, the entire experiment will be described briefly.

Subjects performed sets of 30 sagittally-symmetric, isokinetic trunk extensions in an Asymmetric Reference Frame (ARF, Marras and Mirka, 1989) connected to a Kin-Com (Chattecx, Chattanooga, TN) dynamometer. These sets were performed at progressively increasing loads, the first two sets at 20% of isometric MVC, the next two sets at 30%MVC, and the last two sets at 40% MVC. These sets of exertions were used to generate the fatigue of the low back muscles. Interleaved within these fatiguing trunk exertions were a series of evaluative tests. The tests, including those that are not part of this paper, are listed below. Only tests listed below with asterisks (**) are discussed in this paper.

1. Borg ratings of perceived exertion (Borg, 1998),
2. **one sagittally-symmetric isometric trunk extension maximum exertion,
3. three sets of 10 uni-planar trunk repositioning test with eyes closed (Kelaher et al, 2004),
4. one set of 10 repetitions of a trunk extension force replication test with eyes closed,
5. **one set of 10 continuous sagittally-symmetric, isokinetic (20^{o} /sec) trunk extension exertions at 30% MVC and 0^{o} trunk asymmetry,
6. one set of 11 free-dynamic asymmetric box lift and lowers from the 25 cm above the floor at an angle of 60^{o} to the left side to a height of 84 cm at an angle of 60^{o} to the right side,

7. one set of 3 trunk extension reaction time tests at 30° of trunk flexion with eyes closed, and
8. **one set of 10 continuous 20° asymmetric isokinetic (20° /sec) trunk extension exertions at 30% MVC and 20° trunk asymmetry.

All eight tests listed above were performed along with two sets of the 30 individual repetitions as described above, comprising a single period of the experiment. The first period (period 0), a baseline of the evaluative tests, was performed at the beginning followed by the period 1 (performed at 20% MVC), period 2 (performed at 30% MVC) and, finally period 3 (performed at 40% MVC). Only one subject lasted through the 40% period, all other subjects fatigued in the 30% period. It should be noted that the only item that increased as time into the experiment increased were the force levels in the repetitive trunk extensions, all of the tests were performed with the same parameters throughout the data collection session. Testing was ended at the end of the cycle where the subject first reported a perceived exertion rating of 8 or higher on the 10-point Borg CR-10 scale. This cutoff was chosen to ensure that the subjects were fatiguing, but not at total exhaustion and to allow the subject to finish that cycle of the experiment. A sample timeline of an experimental cycle is provided in Table 1B to help clarify the flow of the experiment.

Table 1B: Timeline of the 20% MVC period.

Task or Test	**Duration (minutes)**
1 set of 30 individual submaximal (20%MVC), sagittally-symmetric, isokinetic (20°/sec) trunk extensions, 0°-50° flexion, 6 repetitions per minute	7
Borg Ratings between 20th-25th exertions of set of 30 trunk extensions	0
**1 isometric trunk extension MVC at 30° trunk flexion, sagittally-symmetric	1
{Subject exits the ARF and gets on test platform}	2
3 sets of 10 trunk repositioning postures, 10 at each of 30° trunk flexion, 30° lateral bending, 30° axial rotation, all with eyes closed	3
{Subject re-enters the ARF}	2
1 set of 10 repetitive submaximal isometric (30% MVC, 30° trunk flexion) force replication exertions, sagittally-symmetric, 1 exertion every 5 seconds	3
**1 set of 10 continuous, submaximal (30%MVC), sagittally-symmetric, isokinetic (20°/sec) trunk extensions, 0°-50° flexion	2
1 set of 30 individual submaximal (20%MVC), sagittally-symmetric, isokinetic (20°/sec) trunk extensions, 0°-50° flexion, 6 repetitions per minute	7
Borg Ratings between 20th-25th exertions of set of 30 trunk extensions	0
1 isometric trunk extension MVC at 30° trunk flexion, sagittally-symmetric	1
{Subject exits the ARF and gets on test platform}	2
1 set of 11 continuous free-dynamic, asymmetric (60° left to 60° right) lift and lower (25cm to 84cm hand height) of weighted milk crate (15 kg)	2
{Subject re-enters the ARF}	2
1 set of 3 reaction time tests at 30° of trunk flexion with eyes closed	2
**1 set of 10 continuous, submaximal (30%MVC), 20° asymmetric (trunk rotated counter-clockwise relative to legs), isokinetic (20°/sec) trunk extensions, 0°-50° flexion	3
TOTAL	**39**

Data Collection and Analysis

Unprocessed electromyography (EMG) was collected and analyzed for tests #5 and #8 above. EMG data was collected from electrodes placed over the following muscles, then saved on a personal computer for further analysis:

- bilateral rectus abdominis (2.5 cm from umbilicus, centered about umbilicus),
- medial external oblique (2.5 cm lateral to palpable edge of rectus abdominis, centered on umbilicus)
- lateral external oblique (lateral most portion of external oblique, centered 2.5 cm superior to umbilicus,
- multifidus (2.5 cm from spine, centered at L4 vertebrae), and
- longissimus (centered 2.5 cm lateral to and 2.5 cm superior to multifidus electrodes).

These unprocessed EMG signals were collected at 1024 Hz and amplified in hardware. The signals were then high-pass filtered at 10 Hz, notch-filtered at 60, 120, and 180 Hz, and full-wave rectified in software. The EMG data was collected for the entire set of 10 repetitions for these tests, then was subsequently averaged over a 2-degree window of 29^{o}-31^{o} of trunk flexion during the concentric portion of each of the 10 separate exertions. The data were then normalized with respect to maximal values obtained during posture-specific (symmetric or asymmetric) isometric trunk extension or flexion MVC exertions with the trunk flexed 30^{o} (Mirka, 1991). Therefore, each EMG data point represents the average muscle activity averaged over a 2^{o} window for a single dynamic concentric exertion. Force, angular velocity, and angle data were collected from the ARF simultaneously with the EMG data and saved to aid in post-processing of the EMG data.

The data were analyzed using ANOVA with a repeated measures experimental design. The independent variables in the analyses were asymmetry ANGLE (0^{o} and 20^{o}) and PERIOD (1, 2, and 3). To simplify the experiment, asymmetry was not randomized during the experiment. This was done to standardize on a single cycle order to simplify the experimental protocol.

RESULTS

Figure 1B shows a decrease in maximal trunk torque output during the isometric maximal trunk extensions that were performed immediately after the fatiguing trunk extension exertions (see Table 1B). This data serves to validate the onset of trunk extensor muscle fatigue in this experimental protocol. An average 15% reduction in maximal torque output was measured from the first maximal exertion in the (0% normalized time to fatigue) to the last maximal exertion before the end of the experiment (100% normalized time to fatigue).

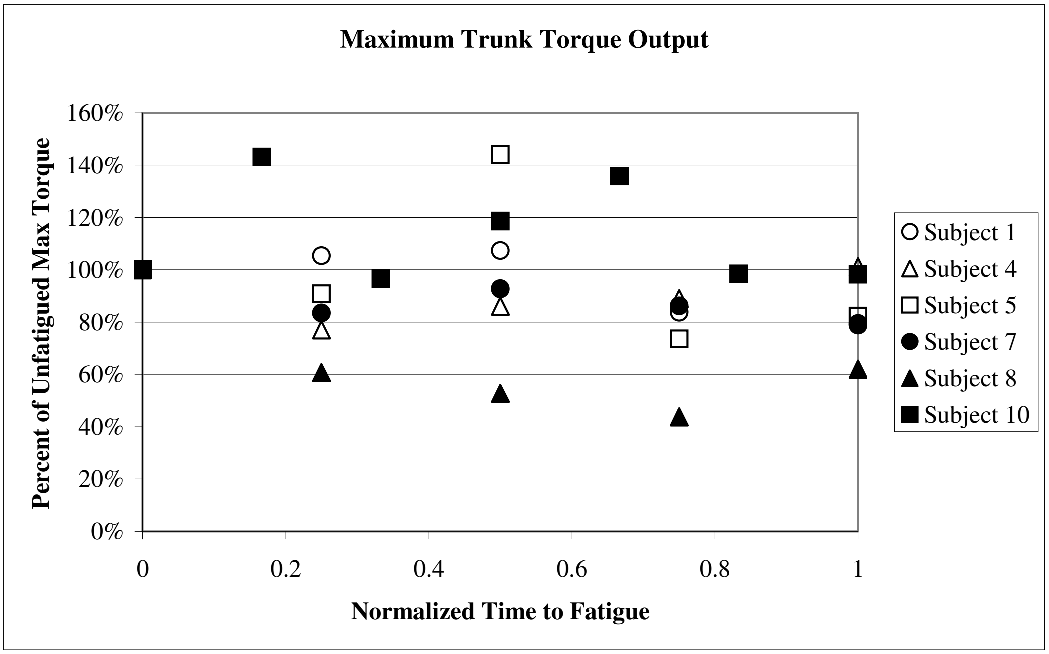

Figure 1B: Scatter plot of individual maximal torque output values as a function of time to fatigue.

One interesting note is that there is not a consistent monotonically decreasing force output continuous over the entire experiment for any individual. This underscores the highly variable nature of the effect of muscle fatigue on maximal torque output within the context of a variable set of tasks.

Figures 2B and 3B show the NEMG of the trunk extensor muscles during the sagittally symmetric and asymmetric isokinetic trunk extension sets (tests #5 and #8), respectively. The statistical analysis showed a significant effect of time PERIOD ($p<0.0001$) showing the well-known increase in NEMG with muscle fatigue with constant force output (Currier, 1969). However, in the symmetric trials (test #5), only the NEMG values for the last time period were significantly different from the others while in the asymmetric trials (test #8) the NEMG for all 3 time periods were significantly different from each other.

The ANOVA on the flexor muscles showed significant main effects of ANGLE ($p<0.0001$), and ANGLE*PERIOD ($p<0.05$). Figure 4B shows that the flexor NEMG values during the symmetric extensions stayed fairly constant throughout the experiment while Figure 5B shows that the flexor NEMG values increased in the asymmetric exertions after Period 2. This visually describes the ANGLE*PERIOD interactive effect. The effect of angle showed a higher mean NEMG for the flexors during the asymmetric extensions than during the sagittally-symmetric extensions (14.4% vs. 11.5%, respectively). Subsequent simple effect ANOVAs showed that the effect of angle was significant at each time period in the experiment.

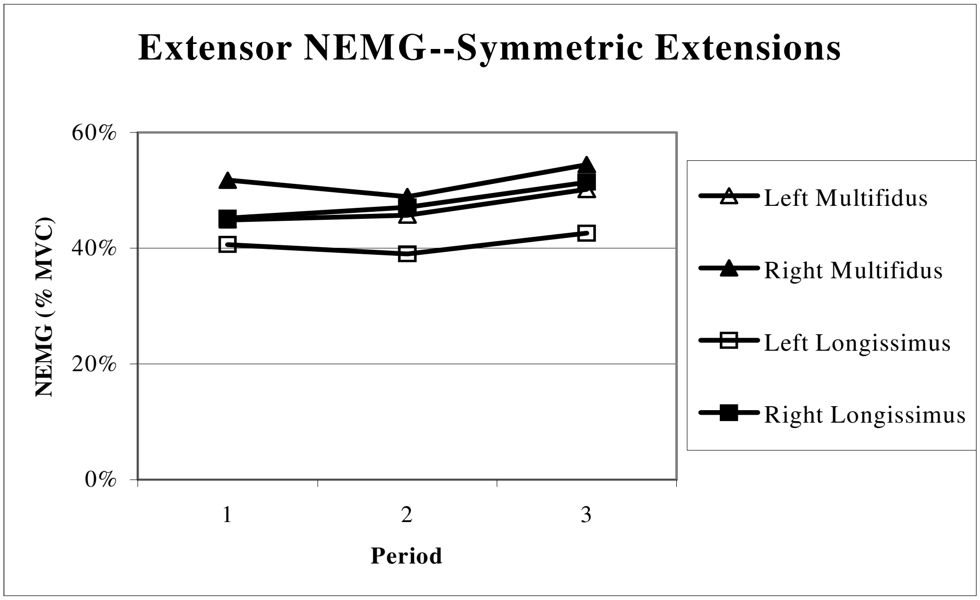

Figure 2B: Normalized EMG (NEMG) of the 4 extensor muscles during the sagittally-symmetric trunk extensions.

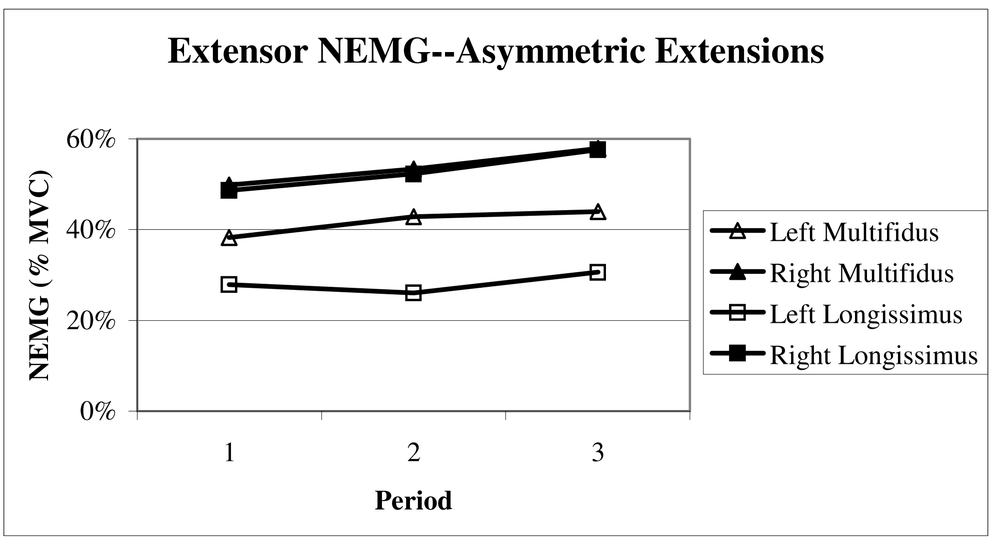

Figure 3B: Normalized EMG (NEMG) of the 4 extensor muscles during the asymmetric trunk extensions.

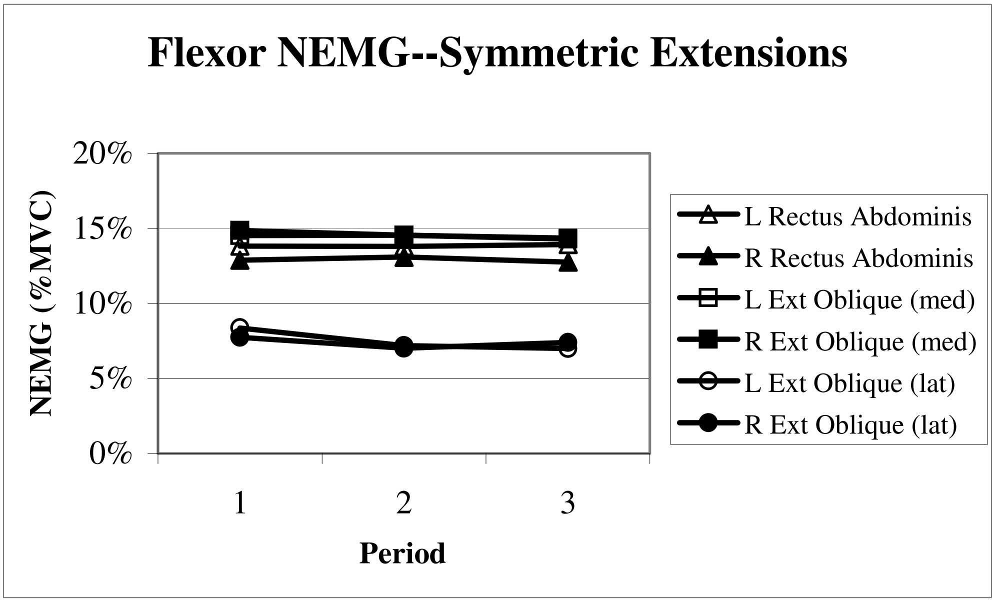

Figure 4B: Normalized EMG (NEMG) of the 6 flexor muscles during the sagittally-symmetric trunk extensions.

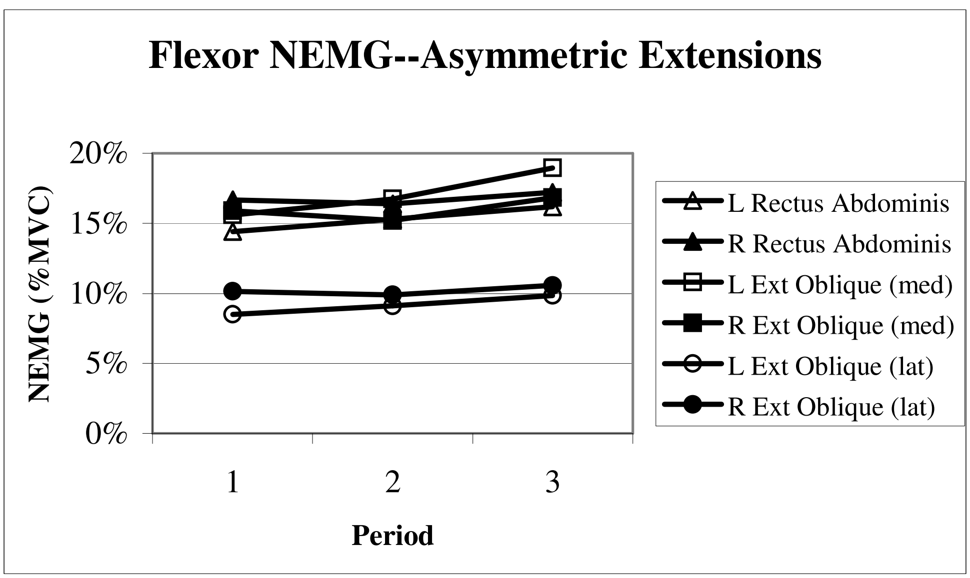

Figure 5B: Normalized EMG (NEMG) of the 6 flexor muscles during the asymmetric trunk extensions.

DISCUSSION

The difference in muscle activity levels of the right and left side muscles during asymmetric trunk extension agrees qualitatively with the data from Dieen (1996). The asymmetry angle causes the right side muscles in this experiment to have a better mechanical advantage than the left side extensor muscles, thus this difference in activation levels is caused by a change in motor pattern to favor the more mechanically advantageous muscles. The differences between Figures 2 and 3 show that fatigue increases the extensor muscle activity levels earlier for the asymmetric trunk exertions then it does for the sagittally-symmetric trunk exertions. The possible cause for this lies in the differences in biomechanical advantage of the left and right side muscle groups. With some fatigue developing in the low back muscles from repeated extension exertions in Period 2, the burden can be distributed to all of the extensor muscles. However, during this same Period for the asymmetric extensions, the slight fatigue by the extensor muscles is enough to require extra recruitment by the right side of the body. In fact, even the left Multifidus muscle in Figure 3 has increased its level of activity in Period 2 before the left Longissimus increases its activity. So it seems that the body may be able to selectively recruit the muscles that have better mechanical advantage first when the extensor muscles start to fatigue.

The increase in NEMG for the flexor data only showed up in the last period of the experiment for the asymmetric extensions. It is possible that this is a reaction to a decrease in stability within the trunk (Granata et al, 2004) due to fatigue in the extensor muscles or a reaction to the more asymmetric loading of the spine due to the asymmetric increases in extensor musculature NEMG. In either case, it can be seen that the effects of trunk extensor muscle fatigue seem to impact asymmetric trunk extensions before they impact sagittally-symmetric trunk extensions. Therefore, it is important to consider this physiological effect of asymmetry during analyses of asymmetric trunk motions in addition to the biomechanical effects of asymmetry. This response to fatigue may be an additional reason why asymmetric lifting tasks are considered a risk factor for occupation low back injuries.

REFERENCES

Borg, G. (1998). *Borg's Perceived Exertion and Pain Scales*. Stockholm: Human Kinetics.

Currier, D.P. (1969). Measurement of muscle fatigue. *Physical Therapy,* 49(7):724-730.

Dieen, J.V.H. (1996) Asymmetry of erector spinae muscle activity in twisted postures and consistency of muscle activation across subjects. *Spine*, 21(22):2651-2661.

Dolan, P. and Adams, M.A. (1998) Repetitive lifting tasks fatigue the back muscles and increase the bending moment acting on the lumbar spine. *Journal of Biomechanics*, 31:713-721.

Granata, K.P., Slota, G.P., and Wilson, S.E. (2004) Influence of fatigue in neuromuscular control of spinal stability. *Human Factors*, 46(1):81-91.

Kelaher, D., Glasscock, N., and Mirka, G. (2004) The effects of body awareness, movement direction, visual condition, and movement speed on trunk proprioception. *Proceedings of the 48th Annual Meeting of the Human Factors and Ergonomics Society.*

Marras, W.S., Lavender, S.A., Leurgans, S.E., Rajulu, S.L., Allread, W.G., Fathallah, F.A., and Ferguson, S.A. (1993) The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders: The effects of workplace factors, trunk position, and trunk motion characteristics on risk of injury. *Spine*, 18(5):617-628.

Marras, W.S. and Mirka, G.A. (1989) Trunk strength during asymmetric trunk motion. *Human Factors*, 31(6), 667-677.

Mirka, G.A. (1991) The quantification of EMG normalization error. *Ergonomics*, 34(3), 343-352.

Parnianpour, M., Nordin M., Kahanovitz N.,and Frankel V. (1988) The triaxial coupling of torque generation of trunk muscles during isometric exertions and the effect of fatiguing isoinertial movements on the motor output and movement patterns. *Spine*, 13(9):982-992.

Potvin, J.R. and O'Brien, P.R. (1998) Trunk muscle co-contraction increases during fatiguing, isometric, lateral bend exertions. Possible implications for spine stability. *Spine*, 23(7):774-780.

Punnet, L., Fine, L.J., Keyserling, W.M., Herrin, G.D., and Chaffin, D.B. (1991) Back disorders and nonneutral trunk postures of automobile assembly workers. *Scandinavian Journal of Work and Environmental Health*, 17:337-346.

METHODS

Experiment Overview

The experimental procedures followed in this experiment are a complicated series of performance assessments interspersed between fatiguing trunk extension exertions. To aid in the understanding of this process, a high-level overview of these procedures is presented first. The detailed description of each component will follow in the subsequent sections. Localized muscle fatigue was generated in six human subjects as they performed sets of 30 sagittally-symmetric isokinetic trunk extension exertions. Interspersed among these sets of fatiguing exertions were a series of "experimental modules" that evaluated various dimensions of the response to the fatiguing exertions. These included: a trunk repositioning module, a trunk extension force replication module, a trunk extensor reaction time module, a three dimensional lifting task module, an EMG-based, sagittally symmetric trunk extension module and an EMG-based, sagittally asymmetric trunk extension module. These six modules were performed before and after the fatiguing exertions to quantify the fatigue-induced changes in each of these measures independently. Each of these tasks were performed multiple times in each module to gain an appreciation for the variability of the response measure at each level of fatigue. (For comparison, a group of three anthropometry-matched control subjects performed the six modules without performing the fatiguing exertions.) Motion analyses of free dynamic asymmetric lifting tasks were performed to analyze the effects of localized muscle fatigue on whole body kinematics. Trunk repositioning tests were performed in each plane of trunk motion to test the effects of localized muscle fatigue on trunk repositioning capability (kinematic effects). Trunk extension force replication tasks were performed to assess the effects of localized muscle fatigue on the force producing portion of proprioception (kinetic effects). Finally, repeated asymmetric and sagittally-symmetric, isokinetic trunk extensions were performed to assess the effects of localized muscle fatigue on muscle coactivation patterns.

To be consistent throughout this entire document, the term "accuracy" will be used to describe average responses in the variables within a set of repeated trials while the term "precision" will be used to describe the variability of the responses within a set of repeated trials. While this may not be standard nomenclature for a description of the average NEMG or average lumbar position, it will be more clear during the results to keep consistent with one term for average response and another single term for variability of a response variable.

Isometric maximal trunk extension exertions and subjective assessments were used to verify the development of fatigue during the experimentation. During the first period of the experiment, only test exertions were performed. During the second period, the subjects performed the fatiguing exertions at 20% of the maximal isometric trunk torque output. During the third period of the experiment, the required torque was increased to 30%. In the final period the torque was increased to 40%. However, only one of the six experimental subjects was able to continue through the 40% period, so data presented in the Results section represents the unfatigued, 20%, and 30% periods. Since the three control subjects did not perform any of the fatiguing exertions, they all finished the experiment by testing through the 40% period. Table 3 shows the timeline of the 20% period.

Table 3: Timeline of the 20% MVC period.

Task or Test	Data Collected	Code*	Duration (minutes)
1 set of 30 submaximal (20%MVC), sagittally-symmetric, isokinetic (20^o/sec) trunk extensions, 0^o-50^o flexion, 6 reps/min	**---**	Fatigue	7
Borg Ratings between 20^{th}-25^{th} exertions of set of 30 trunk extensions	**RPE Ratings**	FV1	0
1 isometric trunk extension MVC at 30^o trunk flexion, sagittally-symmetric	**Extension Force**	FV2	1
{Subject exits the ARF and gets on test platform}	**---**		2
3 sets of 10 trunk repositioning postures, 10 at each of 30^o trunk flexion, 30^o lateral bending, 30^o axial rotation, all with eyes closed	**MotionStar data**	EM1	3
{Subject re-enters the ARF}	**---**		2
1 set of 10 repetitive submaximal isometric (30% MVC, 30^o trunk flexion) force replication exertions, sagittally-symmetric, 1 exertion every 5 seconds	**---**	EM2	3
1 set of 10 continuous, submaximal (30%MVC), sagittally-symmetric, isokinetic (20^o/sec) trunk extensions, 0^o-50^o flexion	**NEMG, ARF force and ARF angle**	EM3	2
1 set of 30 submaximal (20%MVC), sagittally-symmetric, isokinetic (20^o/sec) trunk extensions, 0^o-50^o flexion, 6 reps/min	**---**	Fatigue	7
Borg Ratings between 20^{th}-25^{th} exertions of set of 30 trunk extensions	**RPE Ratings**	FV1	0
1 isometric trunk extension MVC at 30^o trunk flexion, sagittally-symmetric	**Extension Force**	FV2	1
{Subject exits the ARF and gets on test platform}	**---**		2
1 set of 11 continuous free-dynamic, asymmetric (60^o left to 60^o right) lift and lower (25cm to 84cm hand height) of weighted milk crate (15 kg)	**MotionStar data**	EM4	2
{Subject re-enters the ARF}	**---**		2
1 set of 3 muscle onset time tests at 30^o of trunk flexion with eyes closed	**NEMG, ARF force and ARF angle**	EM5	2
1 set of 10 continuous, submaximal (30%MVC), 20^o asymmetric, isokinetic (20^o/sec) trunk extensions, 0^o-50^o flexion	**NEMG, ARF force and ARF angle**	EM6	3
TOTAL			**39**

*Note: Code refers to the identifiers provided later in this section. "Fatigue" represents the fatiguing trunk extension task, FV1 and FV2 represent the Fatigue Verification tests, and EM1-EM6 represent Experimental Modules 1-6.

It should be noted that the tasks that the subjects were asked to perform did require some training to decrease the chance of learning effects confounding the data. To achieve this, two training sessions were provided for each subject to teach them the proprioception, lifting, and dynamometer tasks. No equipment was used in these training sessions in order to minimize the duration of the sessions. Table 4 shows the timeline of the training sessions. The length of each training session was just over 1.5 hours. During one of these practice sessions, a survey was administered to the subjects. A combination of the Body Awareness Questionnaire (BAQ; Shields et al, 1989) and the Kinesthetic and Proprioceptive Awareness Questionnaire (KPAQ; Smith-Jackson, 1998 and Glasscock, 2003) was administered and their scores were added together to gather information on how well the subjects know their own physiology and proprioception capabilities. The actual questionnaire and the results are provided in Appendix A. The total length of one data collection session was just over 4 hours. A complete task timeline for a full data collection session is shown in Appendix B.

Table 4: Timeline of the training sessions.

	Practice Session	**Session 1 (min:sec)**	**Session 2 (min:sec)**
1	Introduction and sign informed consent form	5:00	5:00
2	Stretch	5:00	5:00
3	Describe study and tasks	5:00	5:00
4	Take relevant anthropometric measurements	15:00	(not performed)
5	Take KPAQ survey	(not performed)	15:00
6	Practice position replication tests	5:00	5:00
7	Get subject into ARF	5:00	5:00
8	Adjust ARF to correct dimensions	5:00	(not performed)
9	Practice ARF exertions (record max values)	20:00	20:00
10	Practice fatiguing lifting task	8:00	8:00
11	Practice free dynamic lifting test	3:00	3:00
12	Practice trunk force replication test	3:00	3:00
13	Practice position replication tests	5:00	5:00
14	Get subject into ARF	5:00	5:00
15	Practice ARF exertions (record max values)	10:00	10:00
16	Done	**99:00**	**94:00**

Subjects

Nine healthy adult male subjects, 18-25 years old participated in this experiment. Subjects had no history of any low back pain that had limited their work or leisure activities. All subjects were physically active but were not elite athletes, exercising on average 2.1 days per week (See Table 5). In order to minimize variability of the data due to large anthropometric differences between subjects, all subjects were in the inter-quartile spread of stature (5’6”-6’0”) and weight (140 lb.-200 lb.). Six of the subjects experienced the entire experiment while the remaining three subjects were control subjects, performing all the assessment tests without performing the fatiguing trunk extension tasks. The anthropometric data for the subjects is shown in Table 5.

Table 5: Subject demographic and anthropometric data.

	Experimental Subjects	**Control Subjects**	**All Subjects mean (SD)**
n	6	3	**9**
Age (years)	21.8	21.7	**21.8 (2.5)**
Height (cm)	177.8	177.0	**177.5 (6.1)**
Weight (kg)	74.3	75.6	**74.7 (7.1)**
Exercise **Frequency** (sessions / week)	2.1	4.8	**3.1 (2.4)**

Instrumentation

Three types of instrumentation were used in this experiment. A MotionStar electromagnetic tracking system (Ascension Technologies, Burlington, VT) was used to measure the motion of the subjects during the free dynamic lifting and trunk repositioning tests. An electromyography (EMG) system was used in conjunction with an Asymmetric Reference Frame to collect: 1) muscle activity during controlled trunk extension exertions, 2) muscle onset times during a sudden loading test, and 3) the trunk extension force replication data. The details of these data collection systems will be described next.

The electromagnetic tracking system consists of a large transmitter placed near the subject and smaller individual sensors placed on the subject's body (Figure 4). These sensors were affixed to the clothes or the skin over the right and left greater trochanter of the femur, the sacrum at S1, the T12 vertebral spinous process, and the C7 vertebral spinous process (Figure 5). From this system, the x, y, and z coordinates as well as the roll, pitch, and yaw of each sensor is available. This data was collected at a frequency of 100 Hz for 120 seconds for the entire trunk repositioning test and free dynamic lifting test.

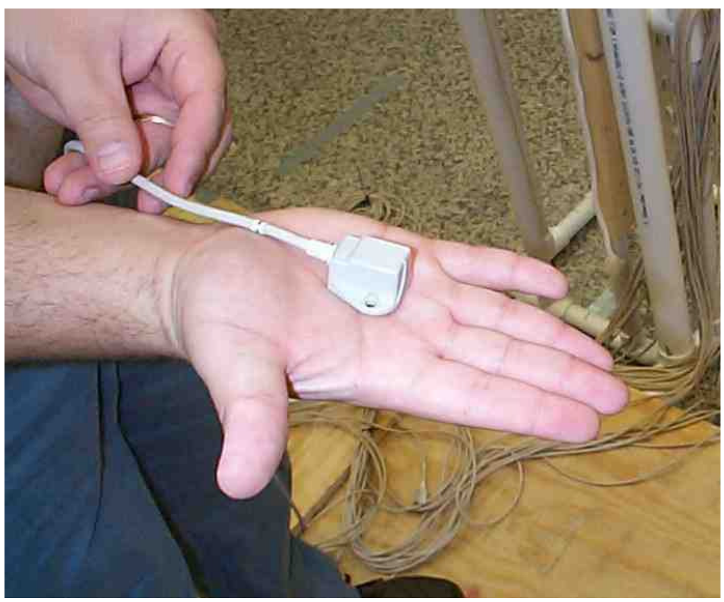

Figure 4: One MotionStar receiver sensor.

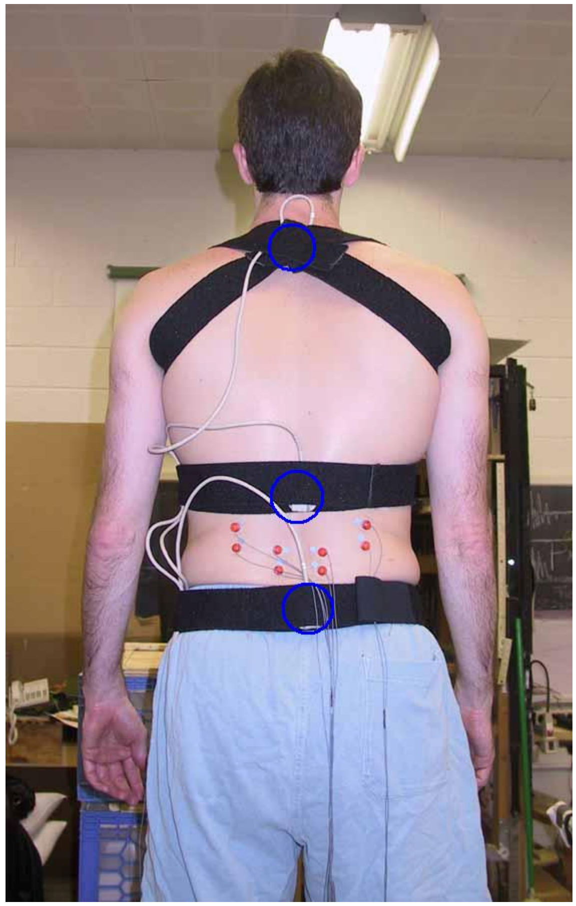

Figure 5: Rear view of subject with Motion Star sensors circled in blue. The sensors are placed underneath the straps to minimize motion during the experiment.

The EMG setup consisted of surface electrodes placed bilaterally over the rectus abdominis, medial external oblique, lateral external oblique, multifidus, and longissimus muscles. These electrodes were placed at an interelectrode distance of 2.5 cm along the lines of the muscle fibers under the electrodes. EMG electrode placements are shown in Figure 6 and followed the placement recommendations of Mirka and Marras (1993). Two external oblique locations were chosen to represent the oblique musculature as a group (Mirka et al, 1997). These EMG signals were notch filtered in post-processing at 60, 120, and 180 Hz to remove electrical noise (and aliases) from the EMG signals. The EMG signals were preamplified at the subject by a factor of 1000 and further amplified at the main data collection system with an adjustable gain of up to 55 times. EMG data was collected at 1024 Hz for each full set of isokinetic trunk extensions in the EMG-based sagittally symmetric and asymmetric trunk extension modules . Full-wave rectified EMG data was normalized for each muscle to the maximal voluntary contraction (MVC) of that muscle, yielding Normalized EMG (NEMG) data reported as %MVC.

Figure 6: Front and rear EMG electrode placement on the subjects.

The Asymmetric Reference Frame (ARF) is a type of exercise apparatus in which the subject stands and can perform planar isokinetic trunk motions and has been used extensively in previous research (e.g., Marras and Mirka, 1992). This frame was used in the fatiguing trunk extension task, the force replication test, the muscle onset time test, the EMG-based sagittally symmetric and asymmetric trunk extension modules. The base of the ARF can rotate to allow for asymmetric lifting motions. This frame is connected to a Kin-Com isokinetic dynamometer (Chattecx, Chattanooga, TN) to proved varied levels of trunk extension velocity and trunk extension torque. The dynamometer force, angle, and angular velocity signals were collected simultaneously with the EMG signals by the data collection computer for analysis at a later time. Also, the subject could view his force output on the screen for feedback during the EMG tests. This was important because the subjects needed to control their torque output during several of the experimental tasks.

Procedures

As illustrated earlier, there was a sequence of experimental tasks performed throughout each period of the experiment. Each of these tasks is described in greater detail below.

Fatiguing Trunk Extension Task

The fatiguing task that was performed was a set of 30 individual isokinetic, concentric/eccentric sagittally-symmetric, trunk extensions in the ARF. The subjects were strapped into the ARF with belts across the top of the thighs and the waist to immobilize the legs and hips during the experiment. The trunk extensions were performed at 20%, 30%, and 40% (if subject had not yet fatigued) of the maximal isometric trunk extension torque (in the second, third, and fourth period, respectively) and all were performed at 20 ^{o}/second. The subjects were instructed to push against the arm of the ARF, which was in contact with their upper back, with a force within 10% of their target force during both the concentric and eccentric portions of the exercise, using visual feedback from the computer monitor connected to the ARF. For example, the target range for the first period was from 18% MVC to 22% MVC, the target range for the second period was 27% MVC to 33% MVC, and the target range for the third period was 36% MVC to 44% MVC. Each extension was started in the upright stationary posture, then the subject flexed his trunk to the 50^{o} trunk flexion posture, then returned to upright standing posture. The subject then rested for four seconds in the upright position after each repetition, then started the next exertion. This yielded a 60% duty cycle (3 seconds exerting concentric force, 3 seconds exerting eccentric force, and 4 seconds resting upright). This was done to simulate individual exertions which are more commonly seen in industry during manual materials handling jobs. The rest breaks also slowed the generation of fatigue, allowing for more data to be collected throughout the experiment. Each set of 30 repetitions took approximately 5 minutes.

The criterion for fatigue was the self-reporting rating of perceived exertion of 8 or higher on any body part during the fatiguing exertions. The ratings of perceived exertion are explained in the following section. It is important to note that the subjects were not tested until exhaustion. This was done to yield data more relevant to occupational tasks where full exhaustion is less likely but a high level of exertion is still being performed.

The principal difference between the procedure followed by the experimental subjects and the procedure followed by the control subjects is this fatiguing trunk extension task component. Whereas the experimental subjects performed the above exertions, the control subjects simply stood in the ARF during this time. The subjects' pelvis was still harnessed in the apparatus, thereby eliminating any real lower extremity mobility, but they did not perform the repetitive trunk extension exertions.

Fatigue Verification 1: Ratings of Perceived Exertion

In order to track the generation of fatigue, subjects rated their perceived effort for different body parts throughout the course of the experiment. This was considered a subjective assessment of fatigue.

<u>Procedure</u>

Starting at the 21st exertion within each set of 30 trunk extension exertions, the subject was asked to provide ratings of perceived exertion (RPE) per the Borg CR-10 scale (Borg, 1998). The subject rated his exertion level for the arms after the 21st exertion, upper back after the 22nd exertion, legs and feet after the 23rd exertion, and the whole body after the 24th exertion. The four second break between subsequent trunk extension exertions was just enough time to allow the subject to rate one body part. Even though the arms and upper back were not stressed much in the experiment, subjects were asked to rate these body parts to identify any whole body discomfort issues were occurring that would not have been evident from the other data.

Fatigue Verification 2: Maximal Trunk Extension

In order to track the effect of fatigue on the trunk extension strength, a single repetition maximum isometric trunk extension exertions were used as verification of the existence of fatigue during the experiment. This was considered an objective assessment of fatigue.

Procedure

One repetition maximum trunk extensions were performed at the beginning of the experiment and immediately after each fatiguing trunk extension task. Immediately after the last trunk extension exertion was completed, the experimenter positioned the subject within the ARF at 30^{o} of trunk flexion and 0^{o} of trunk rotation. The subjects then exerted a maximal isometric trunk extension exertion with their eyes closed, ramping up to maximal force within 1-2 seconds and holding the force for 1 second, then relaxing. The force from the ARF load cells in the arm of the ARF were collected and saved on the computer for analysis after the experiment.

Experimental Module 1: Trunk Repositioning Tests

The effect of repetitive trunk extensions on the proprioception of the trunk was analyzed with a trunk repositioning test. The test performed in this experiment was similar to those performed in prior research (Gill and Callaghan, 1998; Swinkels and Dolan, 2000).

Procedure

This testing was performed on the wooden test platform with the subject secured in a pelvis fixation device (see Figure 7). The wooden platform was necessary to raise the subjects off the floor to reduce the interference between the rebar in the floor and the electromagnetic waves of the electromagnetic tracking system. During the practice session, the three target trunk angles of 30^{o} flexion, 30^{o} lateral bending to the right, and 30^{o} axial rotation to the right were determined with the use of a standard goniometer with a bubble level attached to one of the arms. From these postures, physical targets were made to quickly replicate these target postures later in the experimental sessions. For the 30^{o} flexed posture, the subject bent over to 30^o from vertical. The subject then repeatedly flexed or extended the trunk slightly with the guidance of the experimenter to align the trunk with the arm of the goniometer. The subject was instructed to let his arms hang freely with the palms of the hands together in front of his body with the fingers extended. Next, the experimenter stacked 8" x 8" sheets of ½" thick wooden boards until the subject's fingertips just touched the top of the stack of wooden boards. For the lateral bending, the subject was asked to bend to the right from the upright stance. The experimenter guided the subject until he achieved a posture of 30^o of lateral bending, again using the goniometer to measure the trunk angle. Once the 30^o posture was met, the subject was asked to let his right arm hang straight down with the fingers extended. Then, the experimenter stacked the wooden boards until the subject's middle fingertip just touched the boards. Finally, the 30^o rotated posture was calibrated with a vertical wooden post fastened to the rear of the pelvis fixation device. This wooden post was positioned 30^o to the right and rear of the subject, as measured from the contact point of the subject's backside on the pelvis fixation device. These posture calibration setups allowed the subject to quickly acquire this target posture at the beginning of a new trunk repositioning. The important thing to highlight is that the subject was able to "recalibrate" his target posture for each test, so any deviation is reflective of short-term error as opposed to long-term proprioceptive changes.

Figure 7: Subject against the wooden pelvis fixation device performing a trunk repositioning task.

Subjects performed sets of 10 repetitions of trunk flexion, lateral bending, and trunk rotation. These tests were performed with the eyes closed. Most research on trunk repositioning tasks has isolated feedback to only proprioception by blindfolding the subjects (Taylor and McCloskey, 1990; Ward et al, 1996; Taimela et al, 1999; Swinkels and Dolan, 2000). Even though Newcomer et al (2000) found no difference in trunk repositioning error between eyes open and eyes closed conditions in their experiment, the subjects in the current study were instructed to keep their eyes closed based on pilot testing described in the *Pilot Work* section. Position and orientation of the S1 and C7 landmarks were measured in the trunk repositioning test using the electromagnetic motion analysis system. This system was chosen based on pilot testing which showed better granularity of trunk position measurements than the lumbar motion monitor.

At the beginning of the experiment and after every 39 minutes during the experiment, the subject performed a series of 10 repeated trunk posture repositioning tasks with his eyes closed. Before each set of 10 trunk postures, the subject established the target trunk angles. For each orthogonal plane of trunk motion, the subject attained the target posture with the eyes open, using the wooden boards (for flexion and lateral bending) or vertical post (for the rotated posture) as guides. Then, the subject replicated this position ten times. Then the subject performed this routine for each of the other two planes of motion. These postures were performed at a pace of 3 seconds per repetition and using a simple metronome to aid the subjects in keeping the appropriate pace. This rate was chosen in order to provide an average trunk angular velocity of approximately 30^{o}/sec, to be consistent with Gill and Callahan (1998).

<u>Experimental Design</u>

This experiment utilized a repeated-measures experimental design. The independent variables are the time period in the experiment (1, 2, and 3). Period 1 includes the baseline tests performed before any of the fatiguing trunk extensions had been performed, Period 2 includes the testing performed after the fatiguing exertions performed at the 20% MVC level, and Period 3 includes the testing performed after the fatiguing exertions performed at the

30% MVC level. There were two dependent measures in this module. The dependent variable describing the accuracy of the subject's performance was the absolute error (AE) of the actual trunk position relative to the target position. The second dependent variable was the deviation from the median value. Previous research on trunk repositioning accuracy have used either measure and sometimes both (McNair and Heine, 1999; Swinkels and Dolan, 2000). The deviation from median value was computed as follows. For the ten trunk repositioning tasks (for each plane) the median of these values is computed. The absolute value of the deviation of each observation from this median value was then computed. This approach is similar to that described relative to the Modified Levene's test (Montgomery, 2005). These data provided a measure of the spread (precision) of the response. These analyses were performed independently for each plane of the body.

Experimental Module 2: Trunk Force Replication Test

Procedure

The trunk force replication test was performed immediately after the trunk repositioning test. After the trunk repositioning test, the subject entered the ARF for the trunk force replication test. The subject was secured into the ARF and the EMG electrodes were reconnected to the preamplifiers. The subject flexed forward to 30^{o} of trunk flexion and the arm of the ARF was locked at that position. With eyes open, the subject was instructed to push back against the arm of the ARF with a force equal to 30% MVC, using the force feedback graph from the monitor connected to the ARF KinCom as a target. The graph was setup to show a band of 27%-33% of the MVC force. With the force maintained within the 27%-33% range, the subject was instructed to close his eyes. The subject then stopped pushing against the ARF (but stayed in the 30^o flexed posture), then pushed back against the ARF arm with a force as close to the original target force as possible. The subject was instructed to hold the force for 2 seconds, and then he was instructed to stop pushing against the ARF while still maintaining the 30^o flexed posture. This was performed 10 times, all with the eyes closed. The most stable 1 second of force for each of the 10 repetitions was used to compute the dependent variables of this module.

Data Processing

The target force was calculated as the average force produced during the first second of the data collection file while the subject still had his eyes open while pushing against the arm of the ARF. In the subsequent trials, the average of the 1-second time period wherein the force was held constant was calculated.

Experimental Design

This experiment utilized a repeated-measures experimental design. The independent variables are the time period in the experiment (1, 2, and 3). There were two dependent measures in this module. The dependent variable describing the accuracy of the subject's performance was the absolute error (AE) of the actual torque relative to the target torque. The second measure considered in this module was deviation from median value. This was computed as follows. For the ten trunk extension torque replication tasks, the median of these values is computed. The absolute value of the deviation of each observation from this median value was then computed. These data provided a measure of the spread (precision) of the response.

Experimental Module 3: Sagittally Symmetric Muscle Activity Test

Procedure

Upon completion of the force replication task, subjects performed a set of 10 continuous isokinetic eccentric/concentric trunk extensions in the ARF throughout the same 50° range of motion used in the fatiguing task. The subjects exerted a force against the load cells on the ARF equal to 30% of the initial maximal extension torque generated in the same sagittally-symmetric posture. One set of 10 sagittally-symmetric concentric/eccentric isokinetic trunk extensions was performed in each period. The subjects gauged their force level using the visual feedback from the computer monitor attached to the ARF system. Subjects were required to keep the force within 10% of the target 30% MVC (i.e., the target range was 27% to 33% of the maximum torque). Each exertion was performed at the angular velocity of 20°/sec and performed 10 times. The EMG of bilateral rectus abdominis, medial external oblique, lateral external oblique, multifidus, and longissimus were collected during these test exertions along with the trunk force, trunk flexion angle, and trunk angular velocity.

Data Processing

The EMG data for all muscles were averaged as the subject passed through the window of 31° to 29° of trunk flexion (concentric range of motion only). Since the angular velocity of the ARF arm was 20°/sec and the data collection rate was 1024 Hz, this typically yielded a set of 102 data points within the data collection window. These average EMG data values were then normalized with respect to the posture-specific maximal EMG values gathered in the beginning of the experiment. This yielded the normalized EMG (NEMG) values for each muscle.

<u>Experimental Design</u>

This experiment utilized a repeated-measures experimental design. The independent variable was the time period of the experiment (1, 2, or 3). There were twenty dependent measures in this module. The dependent variables describing the average performance were simply the average (as calculated above) for each of the ten trunk muscles. The second measure considered in this module was, again, the deviation from median value. This was computed as follows. For each of the ten trunk extension exertions the median NEMG for each of the ten trunk muscles was computed. For each of the ten repetitions, the absolute value of the deviation of each NEMG value from this median value was then computed for each muscle. These data provided a measure of the spread (precision) of the EMG response for each muscle.

Experimental Module 4: Trunk Kinematics Test

<u>Procedure</u>

After completing the sagittally symmetric muscle activity test (Module 3) the subject then: 1) repeated the 30 repetition fatiguing trunk extension exertion task, 2) the fatigue verification 1 task, and 2) the fatigue verification 2 task. After completing this task, the subject was released from the ARF and moved onto the wooden platform for the free-dynamic lifting task. In this test the subject was asked to continuously lift and lower a weighted milk crate from a low position to the left of the subject to a high position to the right side. The origin height of 25 cm (10 inches) was located at a 60^{o} angle to the left side of the subject, and the target height was 84 cm (33 inches) located at a 60^{o} angle to the right of the subject. Subjects lifted the milk crate from the low position to the high position, returned to an upright standing posture facing forward for a moment, then lowered the box from the high position to the low position. See Figure 8 for a schematic drawing of the box positions in the lifting task.

The milk crate weighed 33 lb. (15 kg) and was lifted at a self-selected rate by each subject. The horizontal distance between the center of the box and the midpoint between the subjects' feet was 20". The subjects were instructed to lift the crate with their own style and speed but to keep their feet stationary during the test. See Appendix C for the exact instructions given to the subjects. This weight and technique is based on research from Genaidy and Asfour (1989) and Potvin (1992).

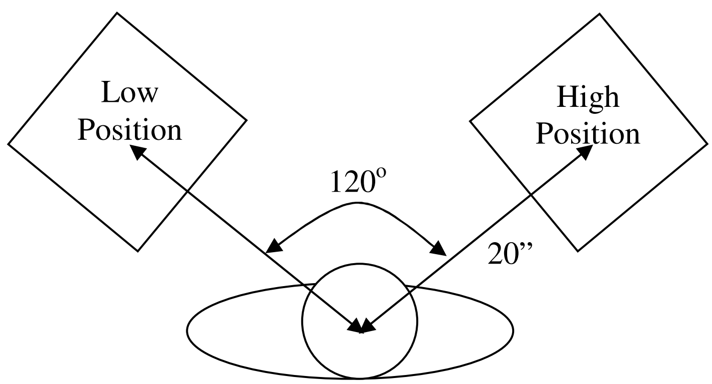

Figure 8: Plan view of the box-lifting task layout.

Figure 9: Isometric of the initial lifting position of the box during the trunk kinematics test.

In this trunk kinematics experiment the electromagnetic tracking system was used to measure the kinematics of the body during the box lifting and lowering test. Ten repetitions of the box lifting and box lowering were measured for whole body kinematics. In addition to the sensors on the bilateral thigh and shank, the S1, T12 and C7 vertebrae, a sensor was placed on the box to track the motion of the box during the lifting task.

Experimental Design

The independent variable for this experiment was the time period in the experiment (1, 2, or 3). The dependent variables describing the average performance were the means of the following variables: 1) lumbar flexion, 2) lumbar lateral bend, 3) lumbar axial rotation, and 4) left and right knee flexion (all computed at the start of the box lift). Also, the maximum angular velocity of the lumbar spine in each of the three planes, and maximum angular acceleration of the lumbar spine in the three planes were computed during the concentric lifting portion of the task as determined from motion of the box in post-processing.

Data Processing

To develop the lumbar flexion, lumbar lateral bend and lumbar axial rotation dependent variables, the difference between the angle of the S1 sensor and the angle of the T12 sensor was calculated in each plane (a similar "relative" procedure was followed to compute the maximum lumbar velocities and maximum lumbar accelerations.) To develop the left and right knee flexion dependent variables, the included angle of the shank and the thigh was calculated. All above data processing was performed using the MotionMonitor software (Innovative Sports Training, Chicago, IL). The specific points in time (i.e. box lift and time at which peak value occurred) were then identified and the dependent values derived.

The second set of dependent measures was the assessment of the variability of these measures between lifts. As with the other modules this was accomplished by considering the deviation from median value. This was computed as follows. For each of the ten concentric lifting motions the median for each of the above kinematics measures was found. For each of the ten repetitions, the absolute value of the deviation of each lift's kinematic measure from this median value was then computed for each muscle. These data provided a measure of the spread (precision) of the kinematic response.

Experimental Module 5: Muscle Onset Time Test

Procedure

After completing the free dynamic lifting module, the subject re-entered the ARF to perform a simple extensor muscle reaction time test. The subject was placed in a sagittally-symmetric posture of 30^{o} trunk flexion and kept his back just touching the arm of the ARF without exerting any force against the arm (this prevented any effects of pre-tensing the muscles on the muscle onset time). The subject was asked to close his eyes for this experiment so he could not predict when to push back against the arm of the ARF. When the subject had his eyes closed with his arms crossed across his chest, the experimenter waited a predetermined number of seconds before he started the motion of the ARF arm. The delay between the keystroke and the start of the motion of the ARF arm was a randomly assigned number between 2 and 8 seconds. When activated, the arm of the ARF started traveling downward at 20^{o}/second, generating a flexion moment about the lumbosacral joint in the subject. The subject was instructed to push back against the arm of the ARF as quickly as he could (see Appendix C for the exact subject instructions for this test). The protocol allowed the subject to go through a range of motion of 10^{o} eccentrically and 10^{o} concentrically before the repetition was over. Each test group was comprised of 3 successive repetitions of the muscle onset time test.

Data Processing

Each repetition of the muscle onset time test consisted of five discrete events that could be identified in the data. These events are listed in below in Table 6. The muscle onset time was calculated as the difference in time between the start of motion of the ARF arm and the onset of muscle activation of each trunk extensor muscle. First, for each trial, the trace of each individual muscle EMG activity (raw signal filtered, full-wave rectified, and averaged over 25 data points) was graphed in a spreadsheet application. Next, the threshold values to determine if muscle activity had started were chosen based on the trace. The method of determining the muscle onset thresholds in previous studies have varied widely and directly impacts the subsequent onset time results. Previous techniques used include visual inspection (Ebig et al, 1997), percentiles of EMG activity (Radebold et al, 2000; mean + 1.4 * SD), initial activity peak (Wilder et al, 1996), and sustained activity above a threshold (DiFabio, 1987). Visual inspection was deemed necessary to choose the thresholds in processing this data since the subjects were positioned in 30^{o} flexed postures at the beginning of the test, thereby reducing the peak to baseline ratio. Formal analysis revealed that objective threshold measurements such as a percentile of the baseline data were not reliable.

Table 6: List of events in a muscle onset time trial.

Event #	Event
1	Experimenter initiates protocol
2	Arm of ARF moves downward towards subject
3	Arm of ARF pushes against subjects' back with force
4	Subject activates his trunk extensor muscles to extend his trunk against the arm of the ARF
5	ARF arm continues downward, causing an eccentric trunk extension exertion through a 10^{o} range of motion

Experimental Module 6: Asymmetric Muscle Activity Test

The sixth experimental module was the asymmetric muscle activity test. All components of this module (Methods, Data Processing and Experimental Design) are identical to those shown in Experimental Module 3 except for the fact that the trunk extension exertions performed by the subjects were done so with the lower extremities rotated 20^o to the right (feet rotated clockwise relative to the shoulders as viewed from above.).

Data Analysis

The analysis of the data collected during this experiment is conducted in two phases, reflecting both the uni-dimensional responses of the measures in the individual modules, as the exploration of the relationships between modules. The principal tool used in the first phase of this analysis is analysis of variance (ANOVA) while correlation techniques will be used to evaluate the relationships between the modules. The details of each of these techniques are described in the following sections.

Statistical Analysis of Individual Modules

Analysis of variance (ANOVA) was the primary evaluation tool used to evaluate the data from each of the individual modules. Prior to conducting this analysis the assumptions of the ANOVA procedure were evaluated using the graphical technique advocated by Montgomery (2005). Where necessary, transformations were applied to the data to address violations of these assumptions (the natural log transformation was found to be appropriate in these data). Examples of this procedure are included in Appendix D of this document. To control for the experiment-wise error rates, adjusted Bonferroni corrections were applied to the p-values used to establish statistical significance. The basic p-value of 0.05 was adjusted based on the number of dependent variables considered in the analysis and the correlations among the dependent measures. A fixed-effect randomized complete block was used to evaluate this data set. Finally, when statistical significance was detected, Tukey's post-hoc test was conducted to further explore the significant relationship. For the fatigue verification components of the research (ratings of perceived exertions and maximum trunk extension force capacity) the primary focus was on the comparison of the responses of the controls vs. the experimental subjects. So, while the experimental modules typically considered only the effect of PERIOD, these fatigue verification modules considered both PERIOD and GROUP as independent variables.

Statistical Analysis of the Inter-Relationships Among the Modules

While the uni-dimensional (module by module) analysis described previously provided some information about how each of these responses was affected by gradual-onset fatigue, an inter-module analysis was necessary to explore the theoretical model developed in this work. The approach used in this study to evaluate these interactions is to explore the correlations between the various dependent variables. Since there are numerous possible results with these six different data sets, all possibilities will not be explored in this research. Those that have biomechanical/physiological plausibility will be the prime focus of this analysis.

The key links in the model are the kinetic links and the kinematic links. These were shown previously in Figure 3. The kinematic links on the left side of the model figure are hypothesized to relate the trunk repositioning tests and the lifting kinematics test. They address the performance of the muscle spindles, which are a primary source of positional information in the trunk. The kinetic links on the right side of the model figure are hypothesized to relate to the muscle activity tests and the trunk force replication test. They relate to the Golgi tendon organs, a primary source of muscle tension regulation in the trunk.

The first kinematic link is the hypothesized relationship between the increased AE of the trunk repositioning tasks (accuracy measures of Module 1) and the changes in the lifting kinematics (accuracy measures of Module 4). This link is supported by consistent changes in these measures in both sets of data. The second kinematic link is the hypothesized relationship between the precision of the trunk repositioning tasks (precision measures of Module 1) and the precision of the lifting kinematics variables obtained during the free dynamic lifting test (precision measures of Module 4). Again, this link is supported with similar trends in the responses of these two measures. The first kinetic link is the suggested relationship between the trunk force replication accuracy (accuracy measures of Module 2) and the accuracy of the EMG data collected during the symmetric and asymmetric trunk extension exertions (accuracy measures of Modules 3 and 6). The second kinetic link is concerned with the trunk force replication precision (precision measures of Module 2) and the precision of the EMG data collected during the symmetric and asymmetric trunk extension exertions (precision measures of Modules 3 and 6). These relationships will be formally evaluated by performing a correlation analysis. A Pearson Correlation Coefficient greater than 0.5 or less than –05 with a p-value <0.05 will be identified as a significant correlation. The relatively conservative threshold for the correlation coefficient was chosen to ensure all interesting relationships were further evaluated, even if the correlation was close to 0.50. A significant correlation will not necessarily indicate a causal relationship, but will lend support to the idea that these variables are physiologically linked together and respond similarly to a fatiguing exertion.

In order to reduce the number of variables in the correlation analysis, several variables were aggregated into logical groups. All of the NEMG variables were grouped into larger categories in this correlation analysis. All of the flexor muscles were summed to create a “flexor” variable for each test and all of the extensor muscles were summed to create an “extensor” variable. All of the values were standardized to the subject-specific unfatigued value from Period 1 by subtracting off the period 1 values from the period 2 and period 3 values. Therefore, every subject’s data has 0 for Period 1 since the first value is always the unfatigued value for all the data.

RESULTS

Verification of Fatigue

The primary reason for the control subjects in this experiment was to ensure that the fatiguing trunk extension task was the cause of any changes found in the dependent variables. Therefore, the results from the control subjects will primarily be discussed in the two fatigue verification tests, the Borg CR-10 perceived exertion ratings and the maximal trunk extension torque test. In each of the subsequent experimental modules, the effects on the control subjects will only be noted if those results are found to be both statistically and biomechanically significant.

Perceived Exertion

The Borg CR-10 ratings for each body part are shown below in Figures 10-14. The graphs show that the experimental subjects felt they were working harder as the experiment progressed, which is expected. Also as expected, the low back perceived exertion ratings were highest among the individual body part ratings. Finally, the consistently low ratings from the control subjects suggest that the long duration of the experiment and the assessment tests themselves provided only modest increases in the perceived exertion by the subjects. There was a significant interactive GROUP*PERIOD effect for the Borg ratings for the whole body ($p<0.01$) and the lower back ($p<0.01$). Also, there was a significant difference between the experimental and control groups for the whole body ($p=0.05$), the legs/feet ($p=0.04$), and the lower back ($p<0.001$). Lastly, there was a significant main effect of time period for the low back Borg ratings (p=0.001). Collectively, these results support the assumption that, from a subjective perspective, the experimental subjects developed significantly higher levels of localized muscle fatigue of the low back than did the control subjects.

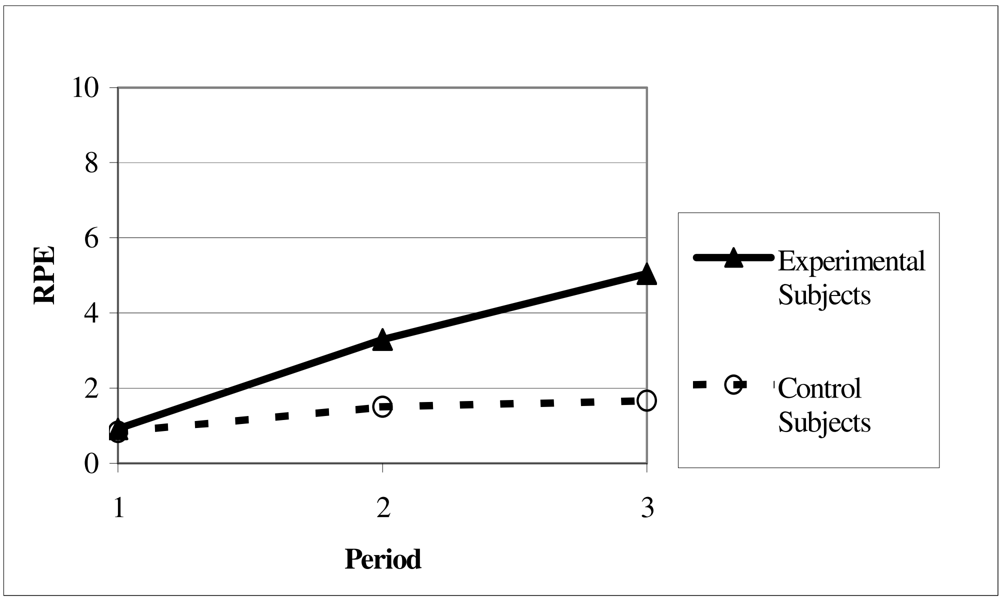

Figure 10: Borg CR-10 ratings of perceived exertion (RPE) for the whole body.

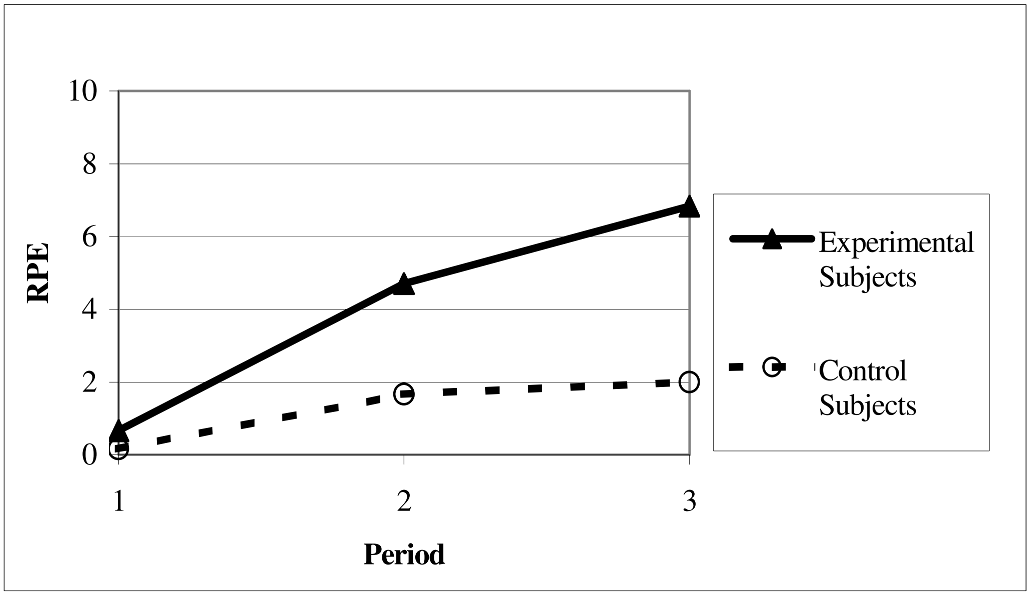

Figure 11: Borg CR-10 RPE for the low back.

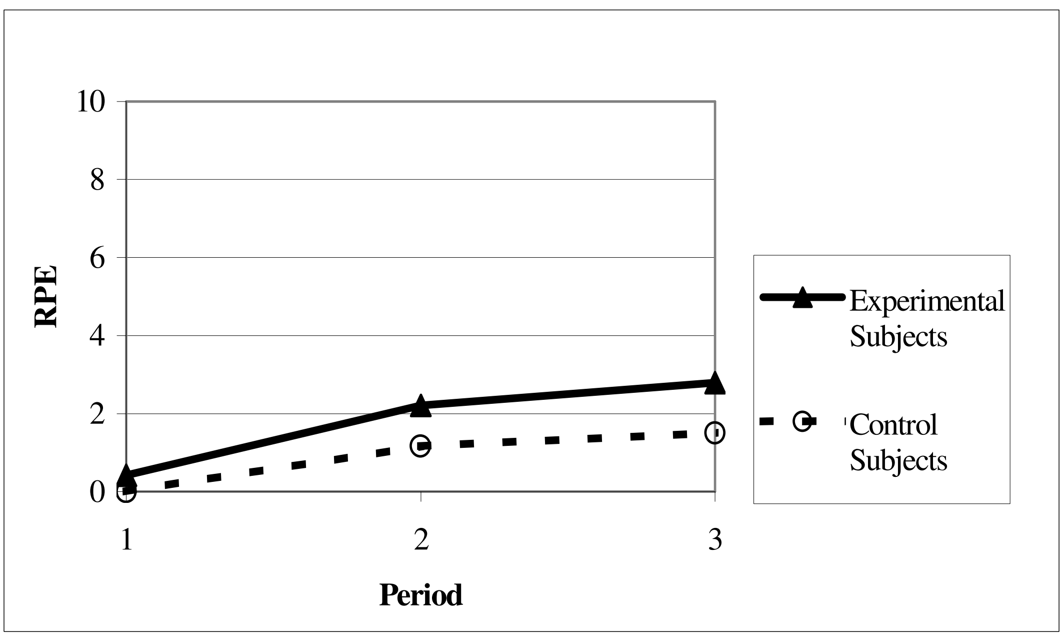

Figure 12: Borg CR-10 ratings for the upper back.

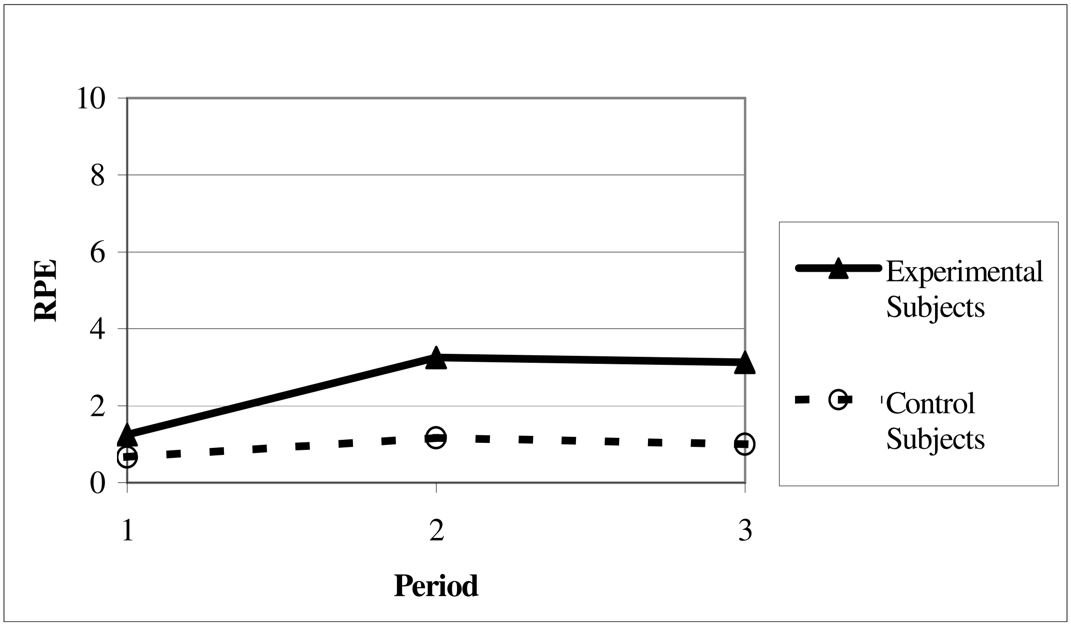

Figure 13: Borg CR-10 RPE for the legs and feet.

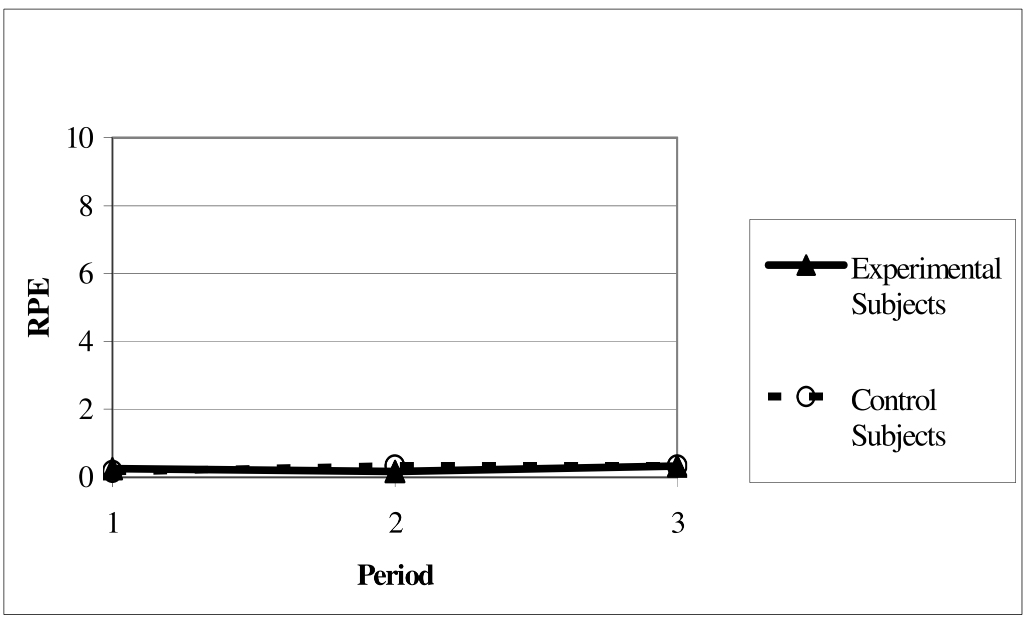

Figure 14: Borg CR-10 RPE for the arms.

Maximal Trunk Extension Torque

Figure 15 and Figure 16 show a decrease in maximal trunk torque output during the isometric maximal trunk extensions that were performed immediately after the fatiguing trunk extension exertions for the experimental subjects (see Table 7). The control subjects showed no consistent trend in maximal trunk extension torque capacity over the experiment. The increase in trunk torque for the control subjects in period 3 may be a learning effect happening over the experiment, though one would expect this to increase monotonically from period 1, which it does not do. This data serves to validate the onset of trunk extensor muscle fatigue in this experimental protocol (Dieen et al, 1996). An average 10% reduction in maximal torque output was measured from the first maximal exertion to the last maximal exertion before the end of the experiment for the experimental subjects, approximately 5% of max torque for each time period. There were no statistically significant main or interactive effects on the absolute trunk extension forces, most likely due to the high individual variability of the force values. However, when the torque values were normalized and expressed as a percentage of each subject's initial maximal torque, there was significant GROUP*PERIOD interactive effect ($p<0.05$). This shows that the maximal torque output of the two groups did respond differently as a function of time into the experiment when the strength of the individuals is taken into account (see Figure 16). Figure 17 shows that the average torque values for both the first and second measurement within each period were similar. Recall that within each period of the experiment, there were two times when the subject was asked to perform the fatiguing exertions. The basic step pattern shown in Figure 17 shows that the level of fatigue at both points in the second period were similar as were the levels of fatigue at both points in the third period. This is important to recognize that all tests conducted in each period can be considered to be at the same level of fatigue.

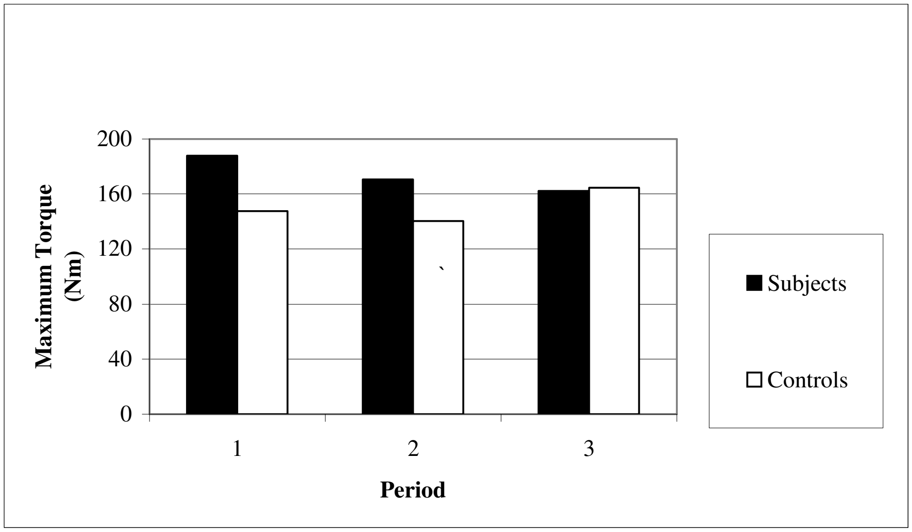

Figure 15: Maximum trunk extension torque values by time and by group.

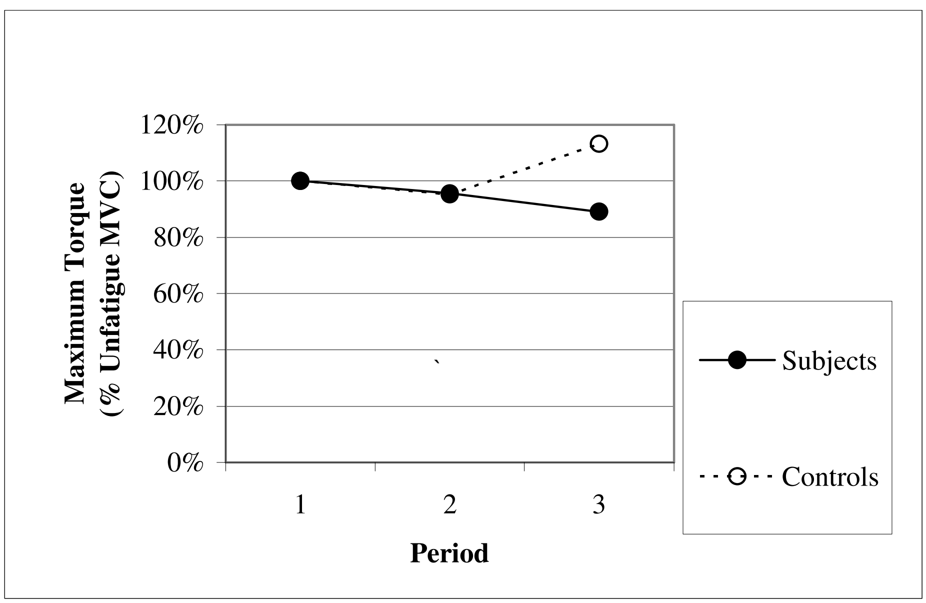

Figure 16: Maximum trunk extension torque values, with each subject's values normalized (100%) to the unfatigued values.

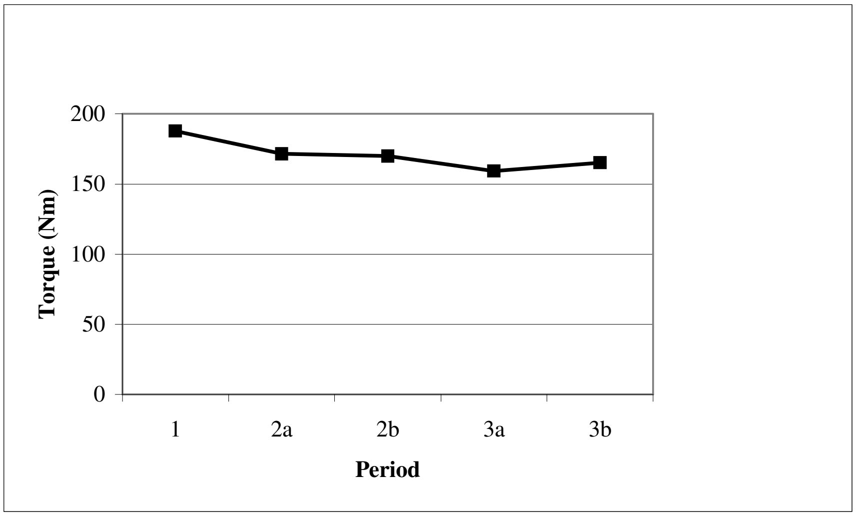

Figure 17: Maximum trunk extension torque values for the experimental subjects showing both sets of values for each time period.

Experimental Module 1: Trunk Repositioning Test

The results of the trunk repositioning tests for the experimental subjects are shown in Table 7 and Table 8. Table 8 shows the ANOVA results for the absolute error (AE) in each plane. There was a significant effect of PERIOD within each plane. Table 9 shows the ANOVA results for the log-transformed precision of the flexion trunk repositioning error with no significant effect of PERIOD. The motion tracking data for one of the experimental subjects had to be discarded because of a corrupted calibration file.

Table 7: ANOVA results of the absolute trunk repositioning accuracy for the experimental subjects (p<0.014 is criterion level)

Plane	PERIOD Effect
Flexion Position Replication Accuracy	**F = 5.54, P = 0.0048*** **
Lateral Position Replication Accuracy	**F = 7.42, P = 0.0009*** **
Axial Position Replication Accuracy	**F = 12.45, P < 0.0001*** **

Table 8: ANOVA results of the trunk repositioning precision for the experimental subjects (p<0.014 is criterion level)

Plane	PERIOD Effect
Flexion Position Replication Precision	F = 0.86, P = 0.43
Lateral Position Replication Precision	F = 3.80, P = 0.025
Axial Position Replication Precision	F = 0.88, P = 0.42

The trunk repositioning data shows an increase in AE for the experimental subjects as a function of PERIOD (Figure 18). Post-hoc Tukey's analysis was performed to identify the significant differences in the AE for the different periods within each plane. In the trunk flexion repositioning test, the AE for period 1 was significantly different from the AE in period 3. In the lateral bending test, the AE for period 3 is significantly different from the AE in periods 1 and 2. Finally, in the axial rotation test, the AE for each period was significantly different from the other two periods. The trunk repositioning precision data did not shown any significant PERIOD effect in any of the three planes.

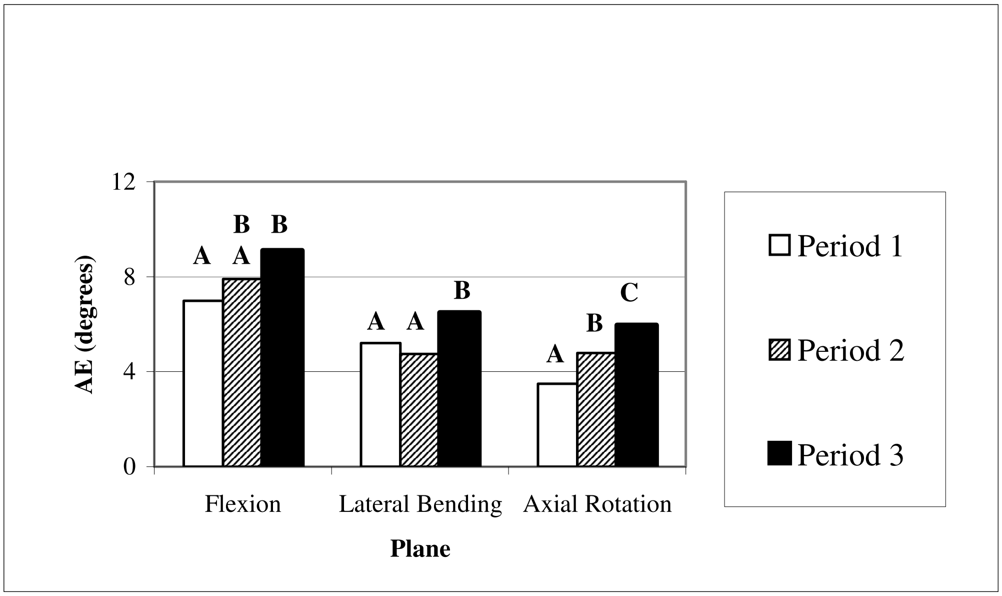

Figure 18: Trunk repositioning AE as a function of PERIOD for the experimental subjects in each of the three planes. Significantly different AE values within each plane are labeled with different letters.

Experimental Module 2: Trunk Force Replication Test

For the experimental subjects, there was a significant effect of PERIOD on the AE but not on the precision of force replication data, as shown in Table 9. Figure 19 shows the AE as a function of PERIOD. The data from one subject in one period was removed from this dataset due to it being an extreme outlier. See Appendix E for the explanation of this decision. The control subjects showed no effect of PERIOD on the accuracy or precision of the force replication test.

Table 9: ANOVA results of the force replication accuracy and precision for the experimental subjects (p<0.025 is the criterion level).

Dependent Measure	PERIOD Effect
Force Replication Accuracy	**F = 9.00, P = 0.0002*** **
Force Replication Precision	F = 0.17, P = 0.85

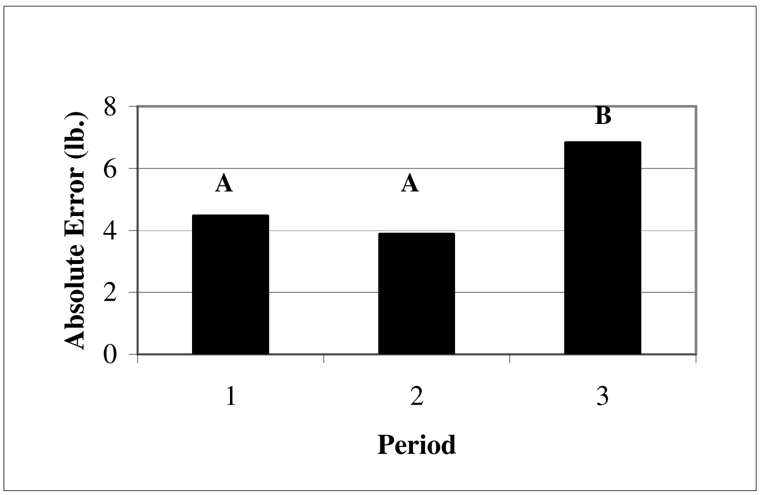

Figure 19: **Absolute force replication accuracy as a function of PERIOD for the experimental subjects. Significantly different values are labeled with different letters.**

Experimental Module 3: Sagittally-Symmetric Muscle Activity Test

The mean muscle activity levels during the sagittally-symmetric muscle activity test show that three of the four trunk extensor muscles were significantly affected by PERIOD, only the left longissimus muscle showed no effect of PERIOD (Table 10). Figure 20 shows the trends for the four extensor muscles across all periods along with the results from the post-hoc Tukey's tests to determine significant within each PERIOD for each muscle.

Table 10: ANOVA results of the NEMG means during the sagittally-symmetric muscle activity test for the experimental subjects (p<0.014 is criterion level).

Muscle	PERIOD Effect
Left Rectus Abdominis	F = 0.45, P = 0.64
Right Rectus Abdominis	F = 0.46, P = 0.63
Left External Oblique (medial)	F = 0.78, P = 0.46
Right External Oblique (medial)	F = 0.57, P = 0.57
Left External Oblique (lateral)	F = 4.13, P = 0.0180
Right External Oblique (lateral)	**F = 6.27, P = 0.0024*** **
Left Multifidus	**F = 13.34, P < 0.0001*** **
Right Multifidus	**F = 10.62, P < 0.0001*** **
Left Longissimus	F = 3.53, P = 0.0315
Right Longissimus	**F = 17.24, P < 0.0001*** **

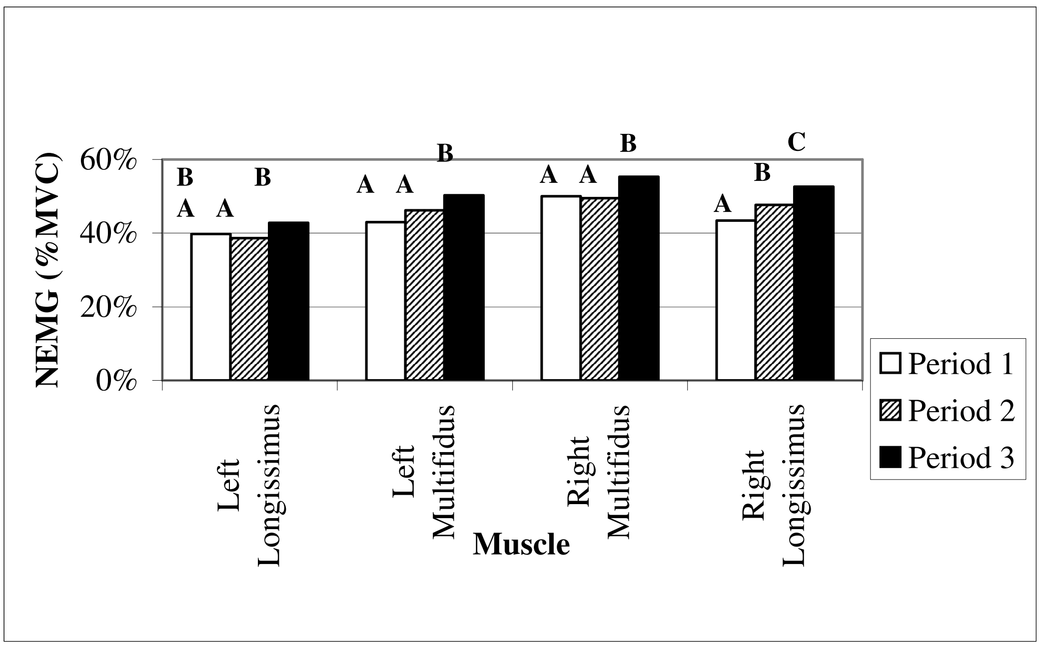

Figure 20: The effect of PERIOD on the extensor NEMG values for the experimental subjects. Significantly different NEMG values within each muscle are labeled with different letters.

Table 11 shows the results of the NEMG variability in the sagittally-symmetric muscle activity test. The trunk flexor data were transformed using the natural log transform due to violations of the ANOVA assumptions while the trunk extensor data was left untransformed as it met the ANOVA assumptions (see Appendix D for graphs testing the ANOVA assumptions). None of the muscles exhibited a significant effect of PERIOD on the NEMG precision for either the control or the experimental group.

Table 11: ANOVA results of the NEMG precision for the sagittally-symmetric muscle activity test for the experimental subjects (p<0.007 is the criterion level).

Muscle	PERIOD Effect
Left Rectus Abdominis	F = 0.14, P = 0.87
Right Rectus Abdominis	F = 0.04, P = 0.96
Left External Oblique (medial)	F = 0.19, P = 0.82
Right External Oblique (medial)	F = 0.19, P = 0.83
Left External Oblique (lateral)	F = 5.18, P = 0.31
Right External Oblique (lateral)	F = 2.53, P = 0.083
Left Multifidus	F = 1.09, P = 0.34
Right Multifidus	F = 0.20, P = 0.82
Left Longissimus	F = 0.10, P = 0.91
Right Longissimus	F = 2.18, P = 0.11

Experimental Module 4: Trunk Kinematics Test

The trunk angles in all three planes and the knee flexion angles at the point of box liftoff all showed a significant effect of PERIOD (Table 12), as shown graphically in Figure 21 through Figure 23. Similarly, all of the lumbar angular velocities in the three planes were statistically significant. The increase in the angular velocity for all planes was approximately 17%. However, none of the angular acceleration data showed any significant effect of

PERIOD. Interestingly the control group showed a significant effect of PERIOD in several dependent variables in this analysis. Lumbar axial rotation at box liftoff (p=.0002), right knee flexion at box liftoff (p=0.0002), and peak lumbar sagittal acceleration (p=0.0057) were all significant (see Figure 24). Neither the controls nor the experimental subjects exhibited any significant effect of PERIOD on the precision of the kinematic data, as shown in Table 13.

Table 12: ANOVA results of the mean trunk kinematics data for the experimental subjects (p<0.009 is the criterion level).

Kinematic Measure	**PERIOD Effect**
Lumbar Flexion at box liftoff	**F = 7.27, P = 0.0010*** **
Lumbar Lateral Bend at box liftoff	**F = 24.98, P < 0.0001*** **
Lumbar Axial Rotation at box liftoff	**F = 60.35, P < 0.0001*** **
Peak Lumbar Sagittal Angular Velocity	**F = 6.96, P = 0.0013*** **
Peak Lumbar Coronal Angular Velocity	**F = 7.55, P < 0.0001*** **
Peak Lumbar Transverse Angular Velocity	**F = 10.01, P < 0.0001*** **
Peak Lumbar Sagittal Angular Acceleration	F = 3.45, P = 0.034
Peak Lumbar Coronal Angular Acceleration	F = 1.62, P = 0.20
Peak Lumbar Transverse Angular Acceleration	F = 1.40, P = 0.25
Left Knee Flexion at box liftoff	**F = 4.79, P = 0.0097*** **
Right Knee Flexion at box liftoff	**F = 6.10, P = 0.0029*** **

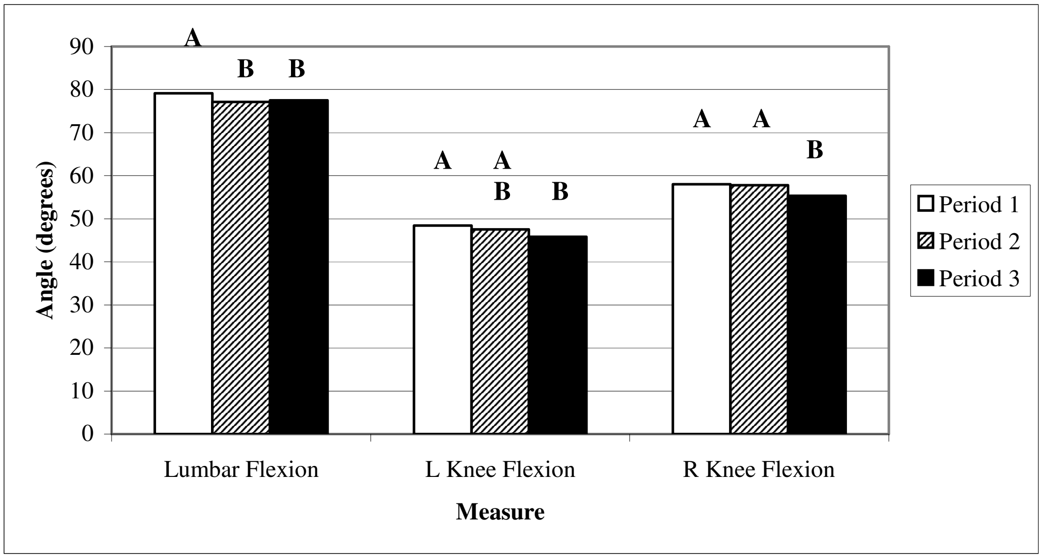

Figure 21: The effect of PERIOD on the kinematic variables for the experimental subjects. Significantly different values are labeled with different letters.

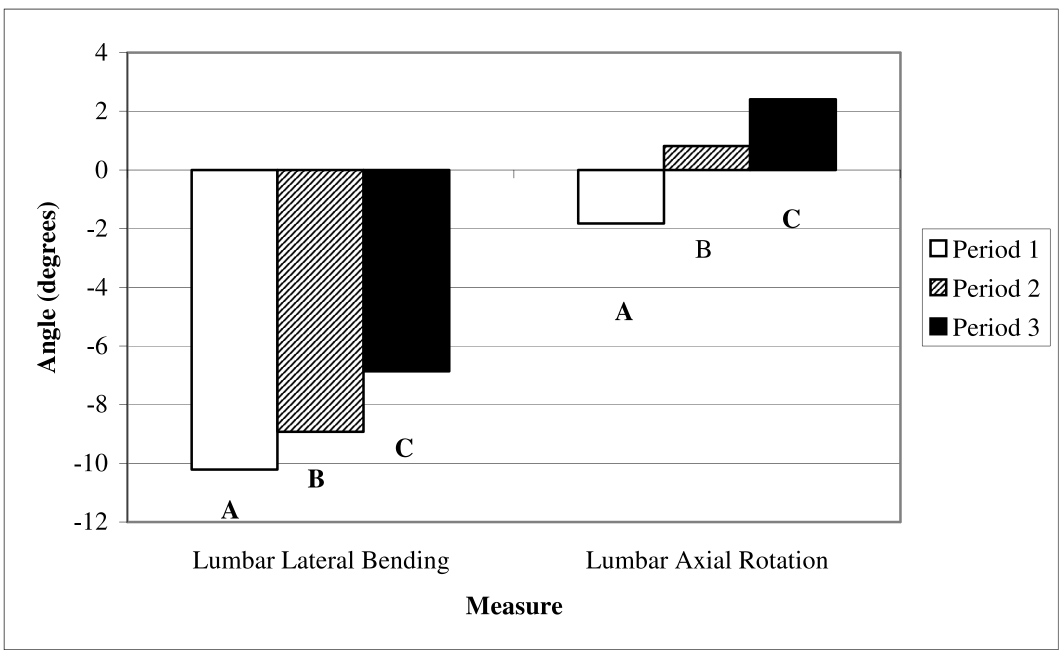

Figure 22: The effect of PERIOD on the kinematic variables for the experimental subjects. Significantly different values are labeled with different letters.

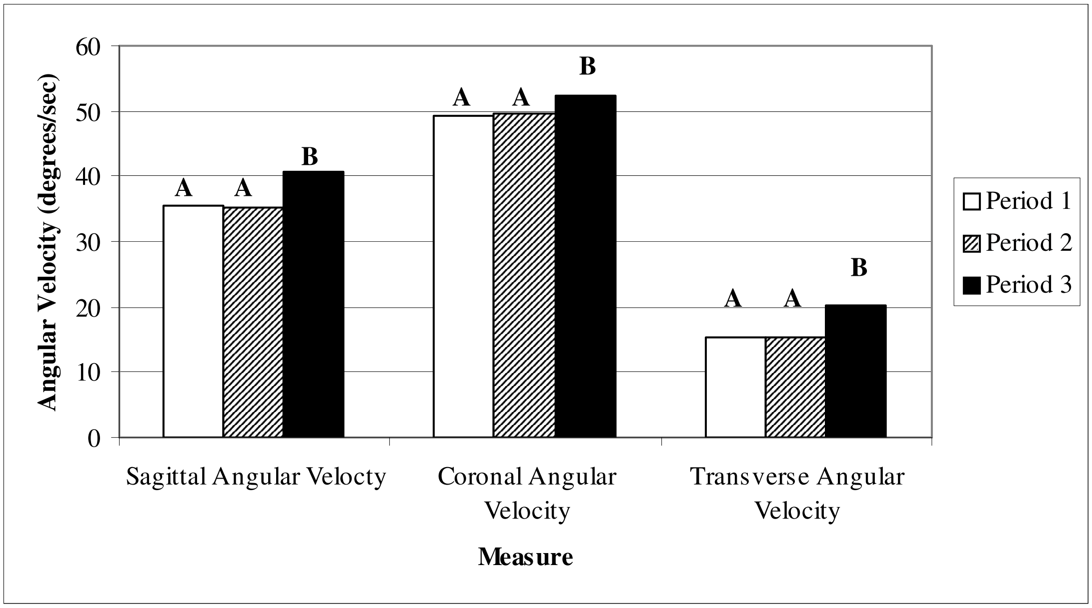

Figure 23: The effect of PERIOD on the kinematic variables for the experimental subjects. Significantly different values are labeled with different letters.

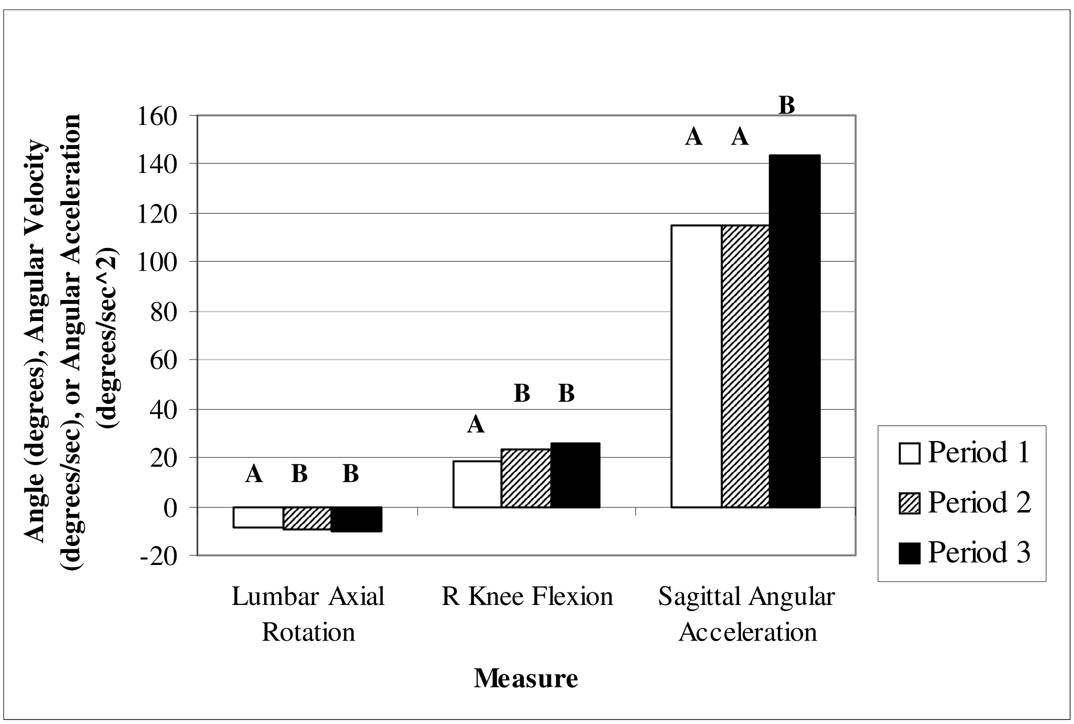

Figure 24: The kinematic variables which exhibited a significant PERIOD effect for the control subjects. Significantly different values are labeled with different letters.

Table 13: ANOVA results of the kinematics precision data for the experimental subjects (p<0.014 is the criterion level).

Kinematic Measure	PERIOD Effect
Lumbar Flexion at box liftoff	F = 1.92, P = 0.15
Lumbar Lateral Bend at box liftoff	F = 0.15, P = 0.86
Lumbar Axial Rotation at box liftoff	F = 0.19, P = 0.83
Peak Lumbar Sagittal Angular Velocity	F = 1.35, P = 0.26
Peak Lumbar Coronal Angular Velocity	F = 0.20, P = 0.82
Peak Lumbar Transverse Angular Velocity	F = 0.50, P = 0.61
Peak Lumbar Sagittal Angular Acceleration	F = 0.28, P = 0.76
Peak Lumbar Coronal Angular Acceleration	F = 0.37, P = 0.69
Peak Lumbar Transverse Angular Acceleration	F = 0.52, P = 0.60
Left Knee Flexion at box liftoff	F = 0.95, P = 0.39
Right Knee Flexion at box liftoff	F = 2.47, P = 0.088

Experimental Module 5: Muscle Onset Time Test

The trunk extensor muscle onset times for both the left longissimus and left multifidus muscles were significantly affected by PERIOD (Table 14). Figure 25 shows the trends of increased muscle onset time (i.e., delay) across all extensor muscles. None of the muscles' onset time precision showed significant effects of PERIOD at the p<0.015 level (Table 15). Finally, the control subjects showed no significant effect of PERIOD on either the mean muscle onset times or the precision of the muscle onset times.

Table 14: ANOVA results of the mean muscle onset times for the experimental subjects (p<0.022 is the criterion level).

Muscle	PERIOD Effect
Left Multifidus	**F = 6.96, P = 0.0023*** **
Right Multifidus	F = 2.49, P = 0.094
Left Longissimus	**F = 4.40, P = 0.0120*** **
Right Longissimus	F = 1.60, P = 0.21

Table 15: ANOVA results of the muscle onset time precision data for the experimental subjects (p<0.015 is the criterion level).

Muscle	PERIOD Effect
Left Multifidus	F = 0.07, P = 0.93
Right Multifidus	F = 1.08, P = 0.35
Left Longissimus	F = 2.02, P = 0.15
Right Longissimus	F = 1.35, P = 0.28

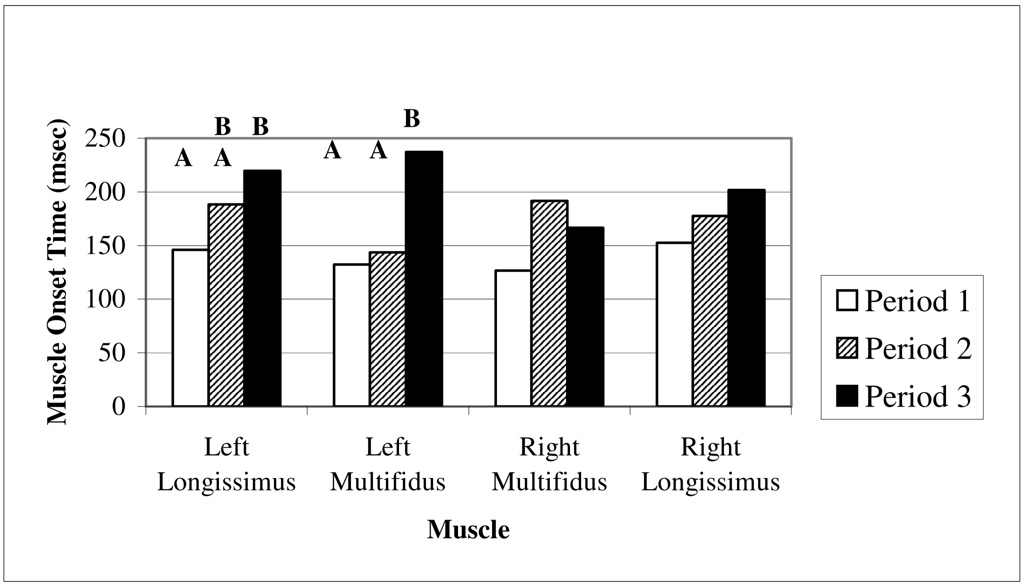

Figure 25: Trunk extensor muscle onset times as a function of PERIOD for the experimental subjects. Significantly different values are labeled with different letters.

Experimental Module 6: Asymmetric Muscle Activity Test

All four trunk extensor muscle NEMG levels were significantly affected by PERIOD in the asymmetric muscle activity test (Table 16). Figure 26 shows the increase in NEMG of the extensor muscles as a function of PERIOD. None of the trunk flexor muscles showed any effect of PERIOD on the mean NEMG values. One muscle (right multifidus) showed a significant ($p<0.0001$) effect for the control subjects (Table 17, Figure 27). Only one muscle (right rectus abdominis) showed a significant effect of PERIOD on the precision of the muscle activity during these asymmetric trials (Figure 28). There were no significant effects of PERIOD on any of the measures of muscle activity precision in the control subjects for the asymmetric muscle activity test.

Table 16: ANOVA results of the NEMG means during the asymmetric muscle activity test. (p<0.014 is the criterion level).

Muscle	PERIOD Effect
Left Rectus Abdominis	F = 1.76, P = 0.18
Right Rectus Abdominis	F = 1.25, P = 0.29
Left External Oblique (medial)	F = 2.28, P = 0.11
Right External Oblique (medial)	F = 2.58, P = 0.08
Left External Oblique (lateral)	F = 0.37, P = 0.69
Right External Oblique (lateral)	F = 4.29, P = 0.016
Left Multifidus	**F = 21.64, P < 0.0001*** **
Right Multifidus	**F = 17.27, P < 0.0001*** **
Left Longissimus	**F = 10.42, P < 0.0001*** **
Right Longissimus	**F = 21.87, P < 0.0001*** **

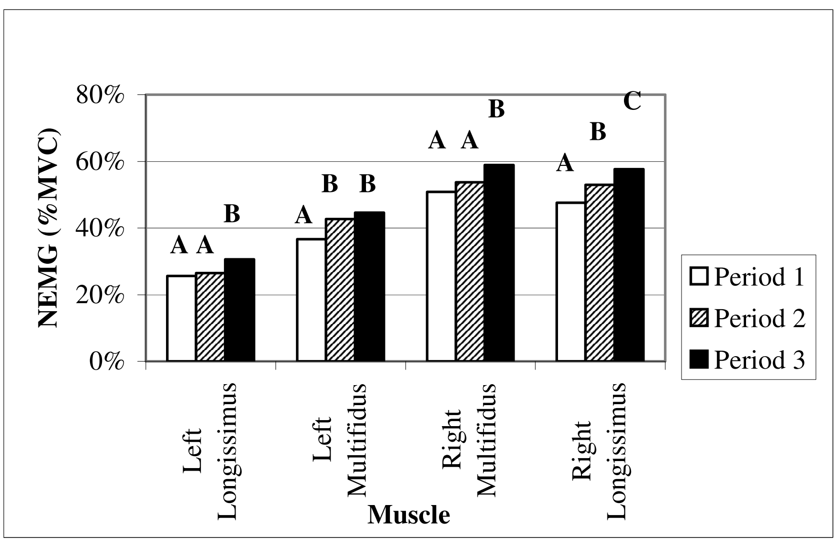

Figure 26: The effect of PERIOD on the extensor NEMG values during the asymmetric muscle activity test for the experimental subjects. Significantly different NEMG values within each muscle are labeled with different letters.

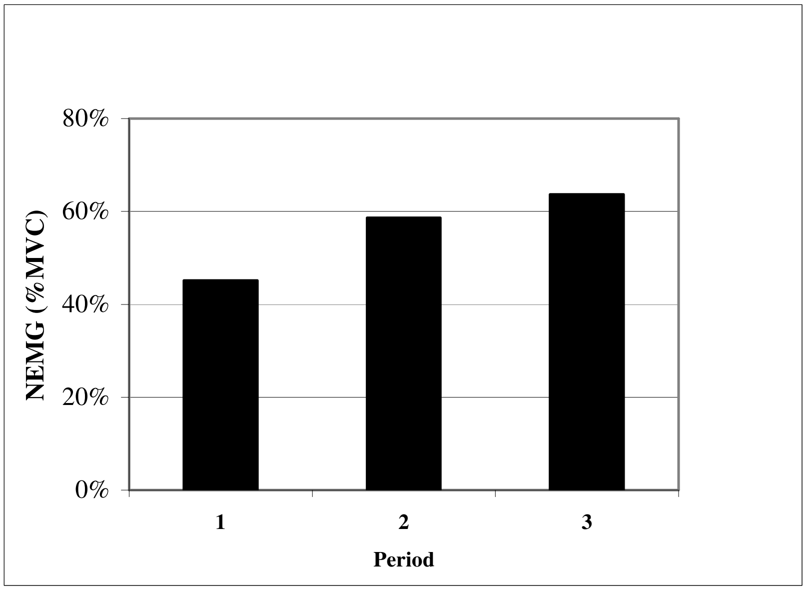

Figure 27: The effect of PERIOD on the Right Multifidus muscle NEMG in the asymmetric muscle activity test for the control subjects.

Table 17: ANOVA results of the NEMG precision for the asymmetric muscle activity test for the experimental subjects (p<0.007 is the criterion level).

Muscle	**PERIOD Effect**
Left Rectus Abdominis	F = 0.96, P = 0.39
Right Rectus Abdominis	**F = 5.43, P = 0.0053*** **
Left External Oblique (medial)	F = 4.54, P = 0.0123
Right External Oblique (medial)	F = 2.76, P = 0.068
Left External Oblique (lateral)	F = 4.39, P = 0.0141
Right External Oblique (lateral)	F = 2.03, P = 0.14
Left Multifidus	F = 1.31, P = 0.27
Right Multifidus	F = 0.15, P = 0.86
Left Longissimus	F = 0.18, P = 0.84
Right Longissimus	F = 0.58, P = 0.56

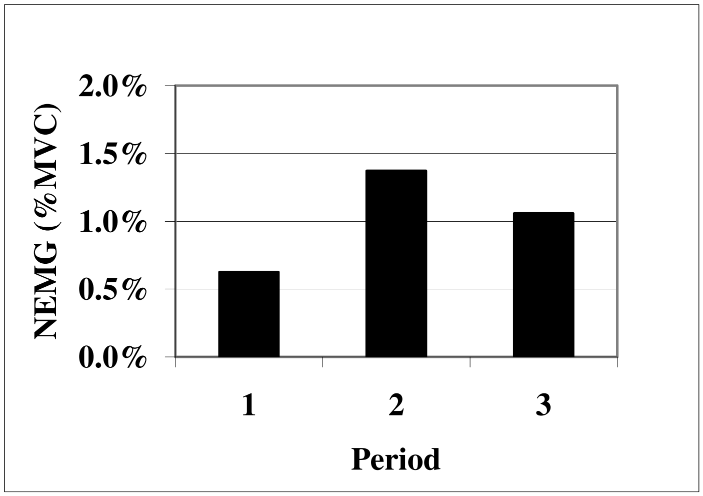

Figure 28: The effect of PERIOD on the rectus abdominis muscle NEMG precision in the asymmetric muscle activity test for the experimental subjects.

Lastly, a summary of the significant effects of PERIOD for the experimental subjects is shown in Table 18. This table shows the how the effect of fatigue strongly impacted the mean accuracy values but not the precision values of the dependent variables.

Table 18: Summary of the significant effects of PERIOD for the experimental subjects.

Test	**Significant effect on Accuracy?**	**Significant effect on Precision?**
Trunk Torque	Yes	NA
Trunk Repositioning	Yes (all)	No
Force Replication	Yes	No
Symmetric Muscle Activity Test	Yes (3/4 extensors, 1/6 flexors)	No
Lifting Kinematics	Yes (angles and velocities)	No
Muscle Onset Time	Yes (left extensors)	No
Asymmetric Muscle Activity Test	Yes (all extensors)	Yes (Right Rectus Abdominis only)

Model Assessment

In an effort to explore the inter-relationships between the response variables considered in the different experimental modules, two types of analyses were conducted. The first analysis was a high-level graphical assessment to identify similarities in trends for each of the response measures. The second analysis was a more structured statistical analysis utilizing correlation procedures. The results of these analyses are shown below.

Graphical Analysis

One can look at the overall trends in the results from the individual tests to get a general sense of the average responses of the experimental group in these various dependent measures. These are all shown in Figure 29 through Figure 41. As previously mentioned, the data in this analysis were all standardized to the unfatigued values by subtracting off the values from Period 1. Therefore, all of the graphs show the deviations in the data from the period 1 data. Raw data graphs for each individual for these measures supplemented these group trends and are provided in Appendix F.

At this superficial level it can be observed that these trends can be placed into one of three categories: not responsive to PERIOD, linearly responsive to PERIOD, and non-linearly responsive to PERIOD. The response variables are all categorized in Table 19. This differentiation of the responses potentially allows for different variables to be related to others in different tests, providing support for the components in the theoretical model. While this graphical analysis generates an overall understanding of these responses it is only when the objective correlation analysis is performed that a better understanding of the these links can be obtained. This was the goal of the correlation analysis.

Table 19: Trend categories of response variables in terms of time period.

Measures not responsive to PERIOD	Measures linearly responsive to PERIOD	Measures non-linearly responsive to PERIOD
Flexor NEMG in symmetric trunk extensions	Extensor NEMG in symmetric trunk extensions	Extensor muscle NEMG in asymmetric trunk extensions
Flexor NEMG precision in symmetric trunk extensions	Extensor NEMG precision in symmetric trunk extensions	Extensor muscle NEMG precision in asymmetric trunk extensions
Flexor NEMG in asymmetric trunk extensions	Flexor NEMG precision in asymmetric trunk extensions	Force replication accuracy
Extensor NEMG precision in asymmetric trunk extensions	Trunk extensor muscle onset time	Trunk repositioning accuracy (lateral bending)
Force replication precision	Maximum trunk extension torque	Trunk kinematics accuracy (flexion, sagittal angular velocity, transverse angular velocity)
Lumbar flexion at box liftoff in kinematic test	Trunk repositioning accuracy (flexion and rotation)	Trunk kinematics precision (sagittal angular velocity, knee flexion)
Trunk kinematics precision (flexion, lateral bending, rotation)	Trunk kinematics accuracy (lateral bending, rotation, coronal angular velocity, knee flexion)	
	Trunk kinematics precision (coronal angular velocity, transverse angular velocity)	

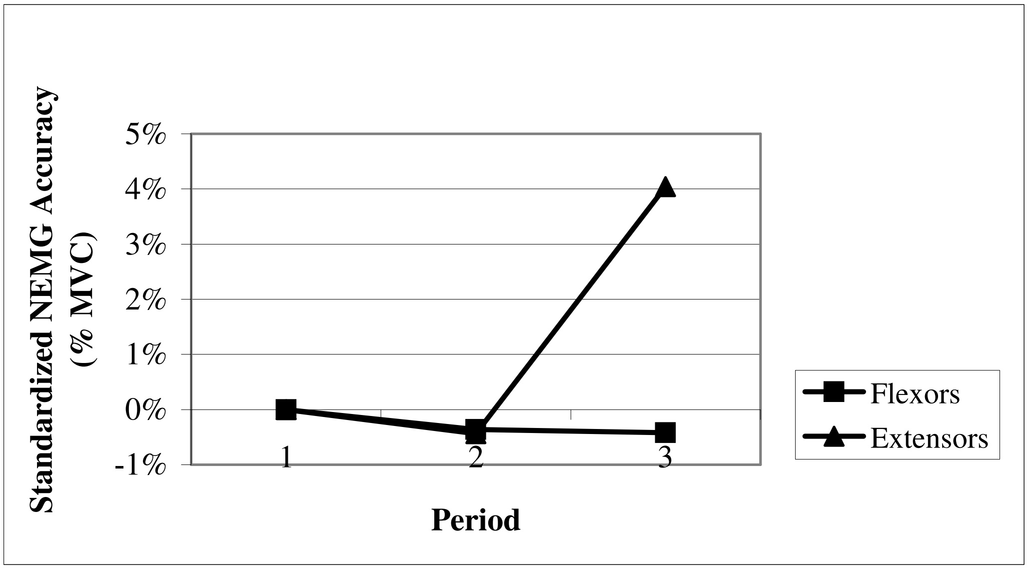

Figure 29: The effect of PERIOD on the mean flexor and extensor NEMG values during the symmetric muscle activity test, standardized to the initial unfatigued values.

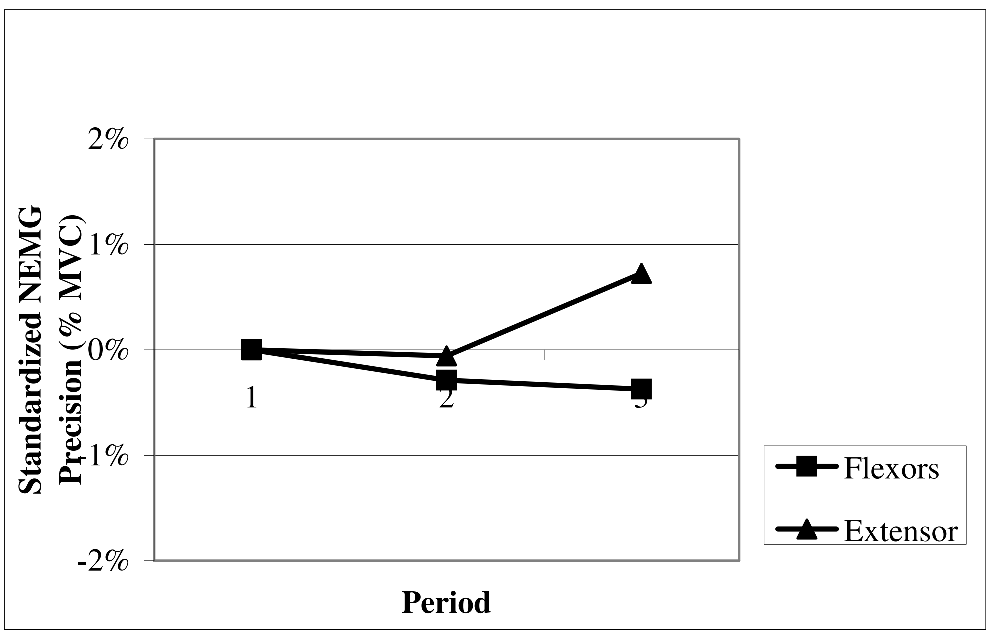

Figure 30: The effect of PERIOD on the flexor and extensor NEMG precision during the symmetric muscle activity test, standardized to the initial unfatigued values.

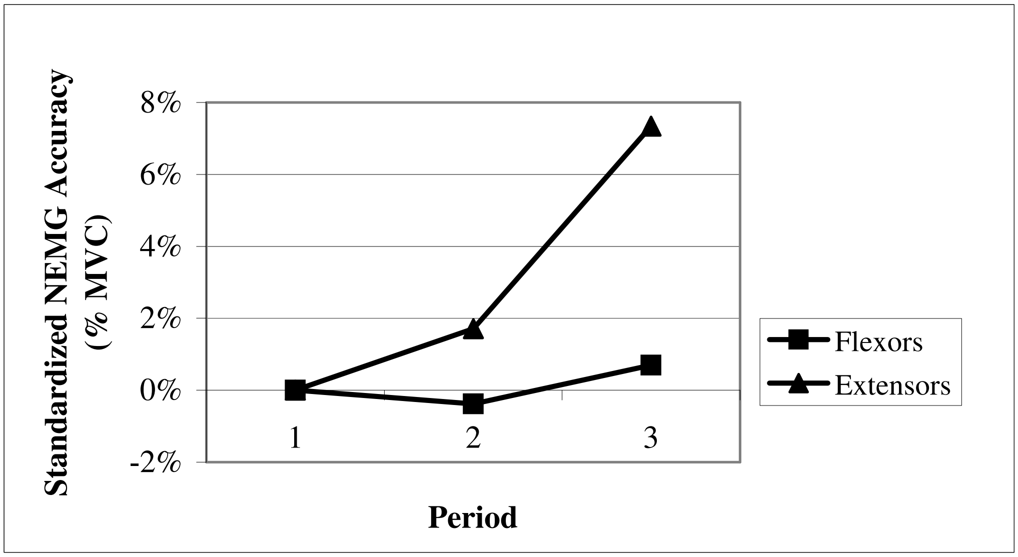

Figure 31: The effect of PERIOD on the flexor and extensor NEMG values during the asymmetric muscle activity test, standardized to the initial unfatigued values.

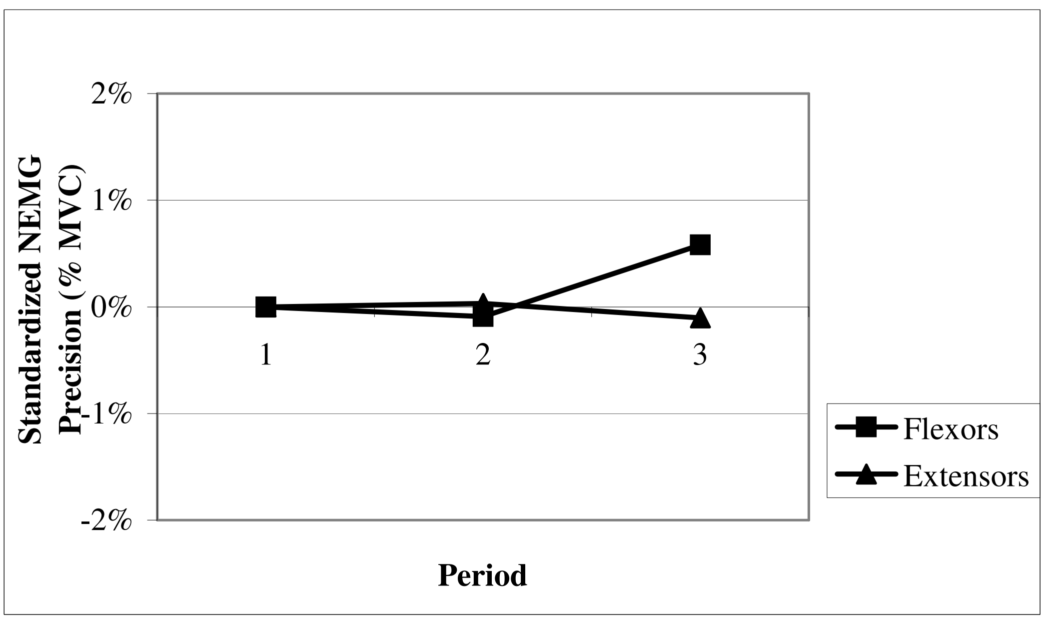

Figure 32: The effect of PERIOD on the flexor and extensor NEMG precision during the asymmetric muscle activity test, standardized to the initial unfatigued values.

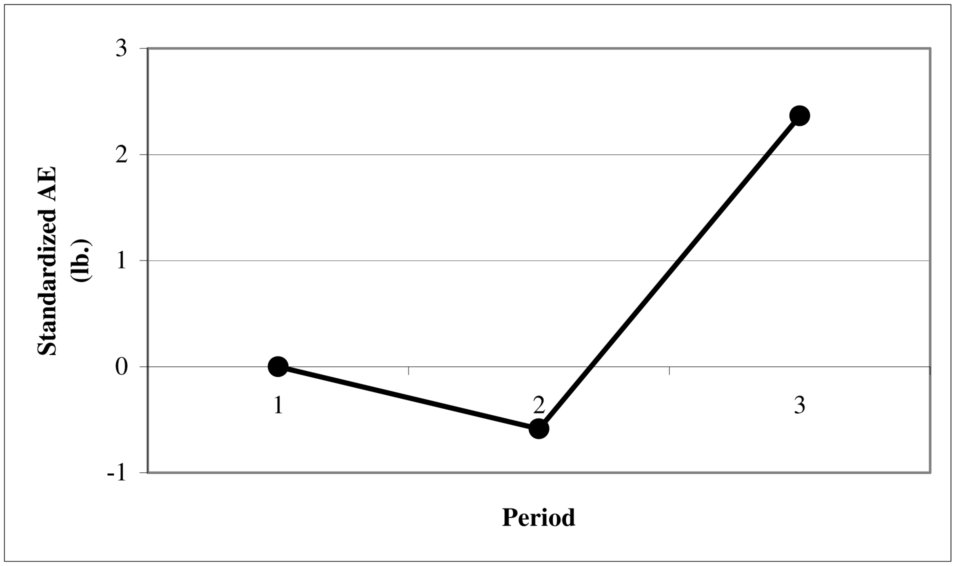

Figure 33: The effect of PERIOD on the force replication accuracy for the experimental subjects, standardized to the initial unfatigued values.

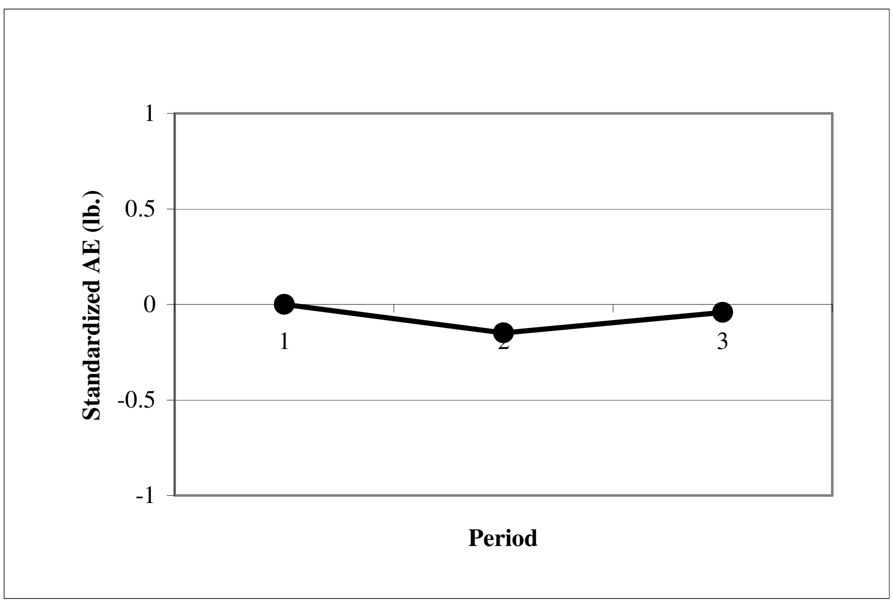

Figure 34: The effect of PERIOD on the force replication precision for the experimental subjects, standardized to the initial unfatigued values.

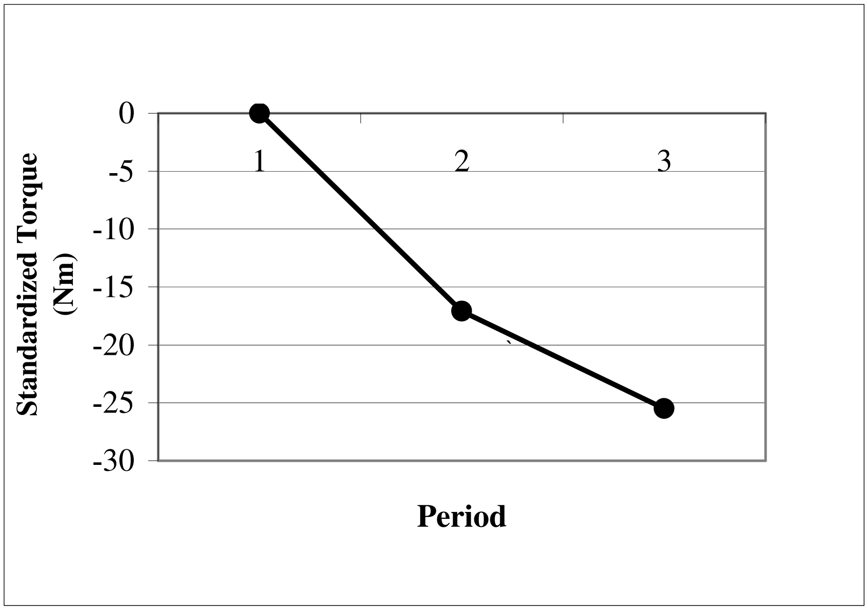

Figure 35: The effect of PERIOD on the maximal trunk torque for the experimental subjects, standardized to the initial unfatigued values.

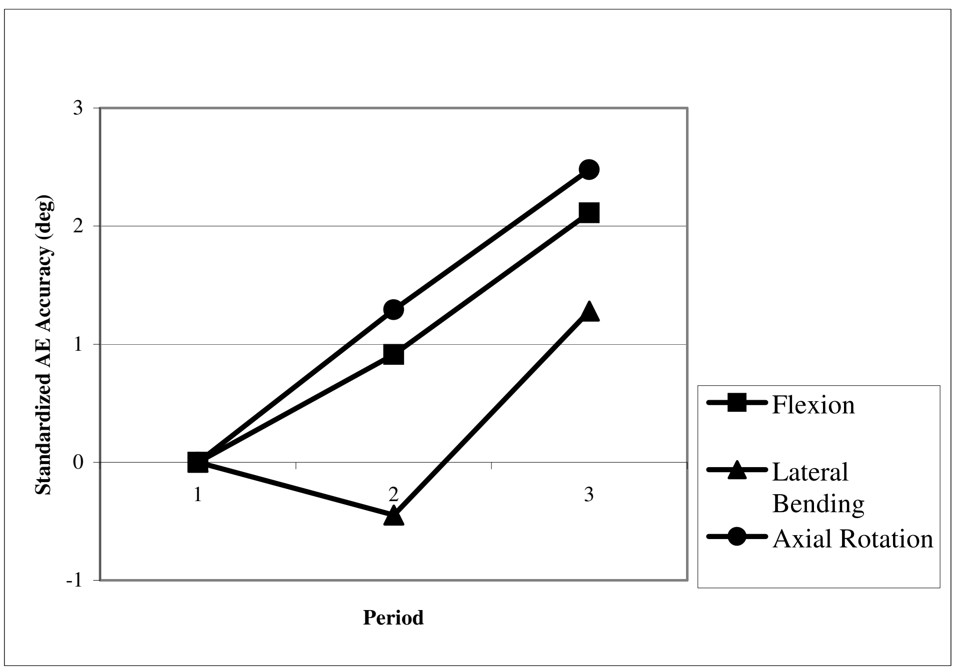

Figure 36: The effect of PERIOD on the trunk repositioning accuracy in the three planes for the experimental subjects, standardized to the initial unfatigued values.

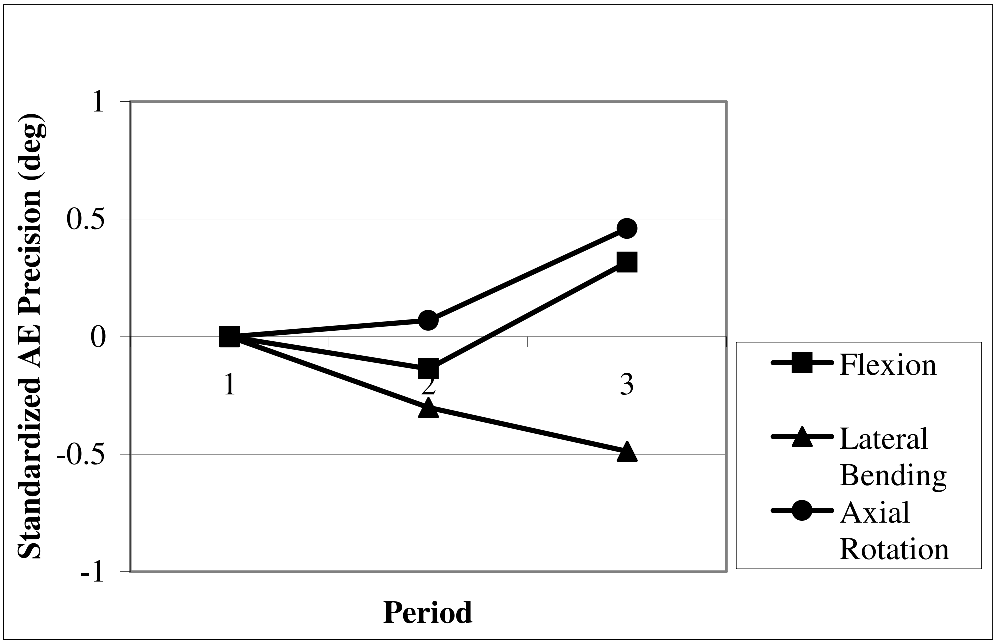

Figure 37: The effect of PERIOD on the trunk repositioning precision in the three planes for the experimental subjects, standardized to the initial unfatigued values.

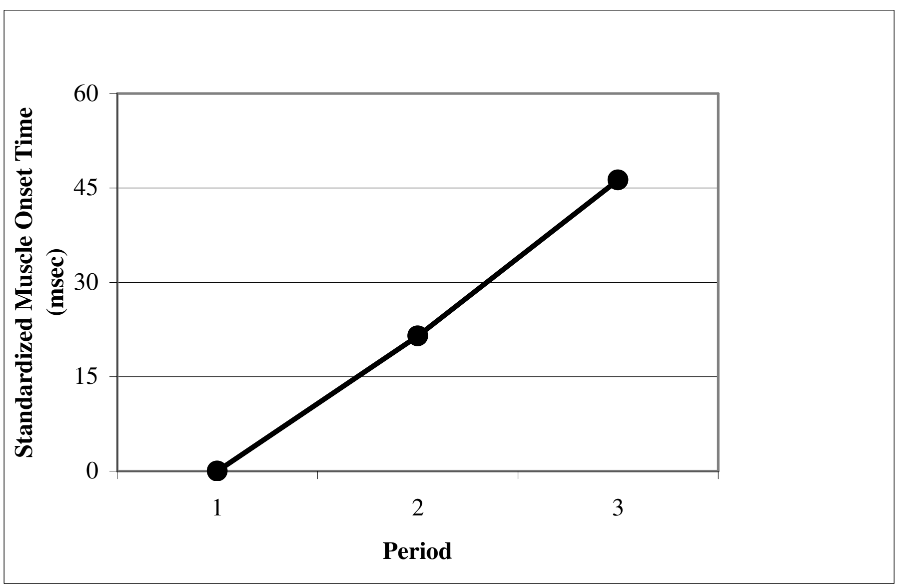

Figure 38: The effect of PERIOD on the average trunk extensor muscle onset times for the experimental subjects, standardized to the initial unfatigued values.

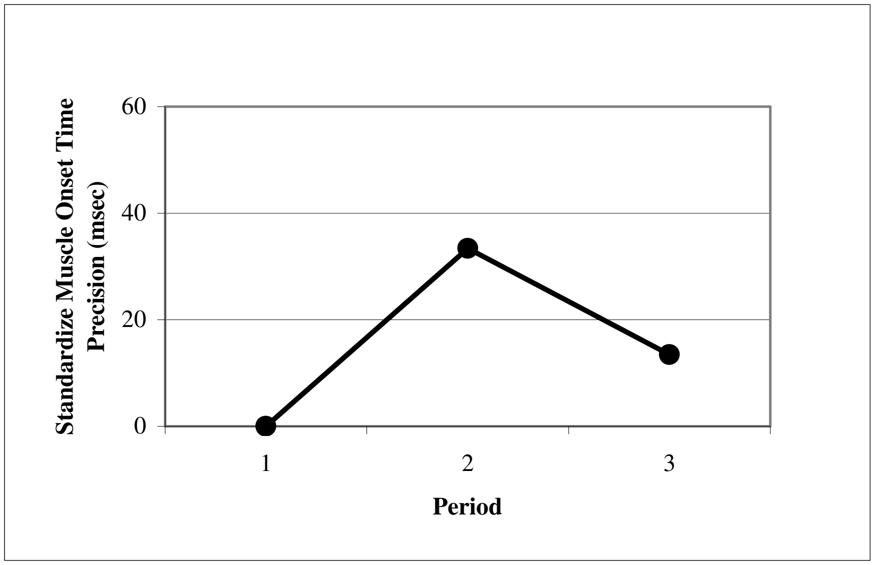

Figure 39: The effect of PERIOD on the trunk extensor muscle onset time precision for the experimental subjects, standardized to the initial unfatigued values.

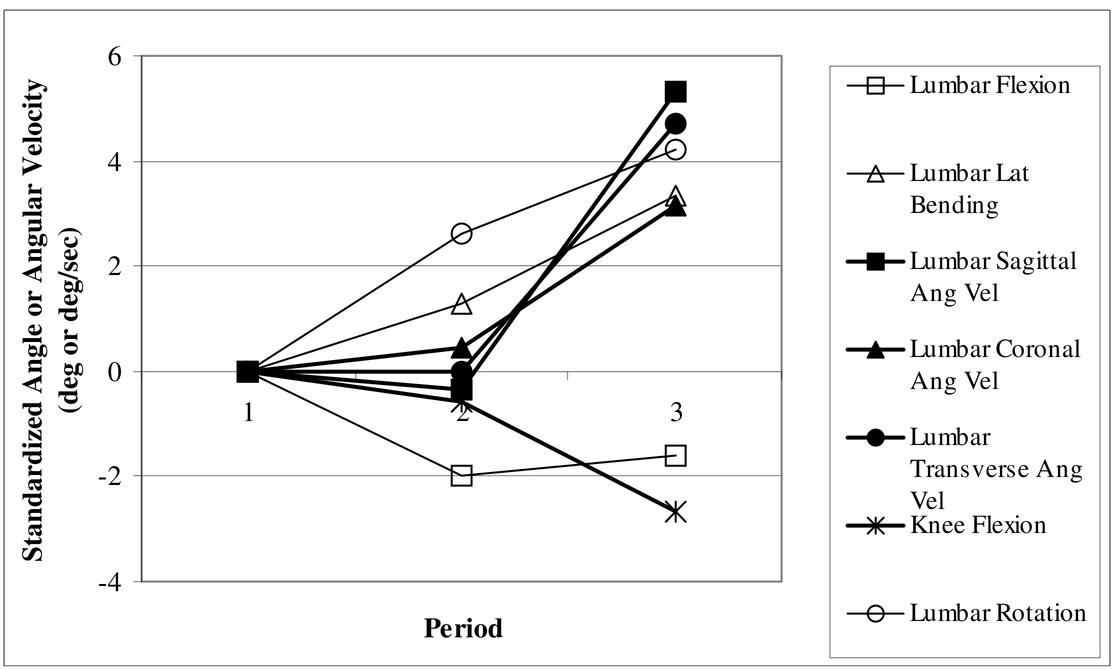

Figure 40: The effect of PERIOD on the accuracy of the angle and angular velocity data from the kinematic test for the experimental subjects, standardized to the initial unfatigued values. The knee angle is the average of the left and right knee flexion angle for each subject.

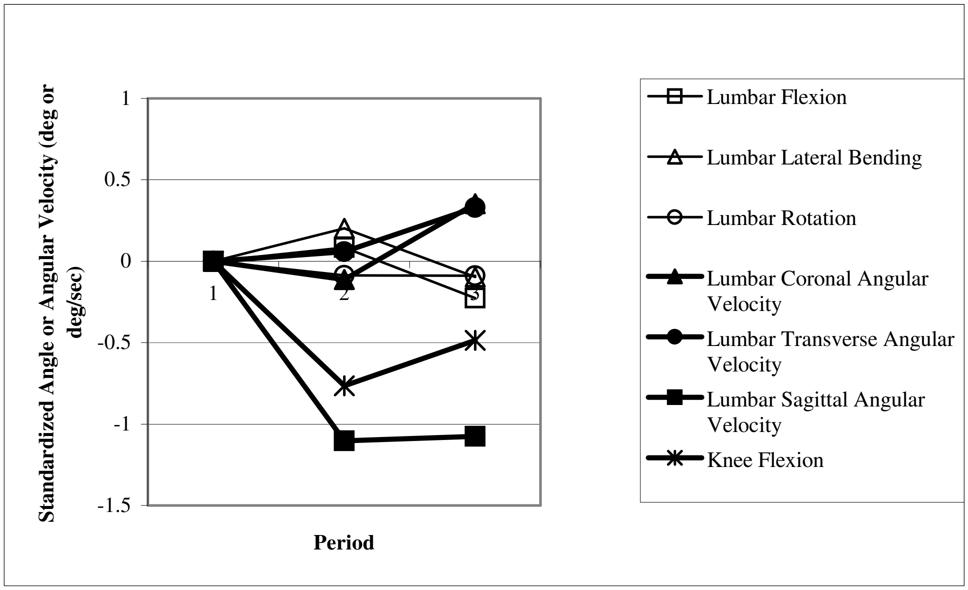

Figure 41: The effect of PERIOD on the precision of the angle and angular velocity data from the kinematic test for the experimental subjects, standardized to the initial unfatigued values. The knee angle is the average of the left and right knee flexion angle for each subject.

Correlation Analysis

The important correlation coefficients in this analysis are the correlations across tests rather than within the same test. These correlations across tests indicates the relationship of the different biomechanical and proprioceptive measures as they react to the fatiguing trunk exercise. These correlations relate to the kinematic and kinetic links that were proposed in the theoretical model section and they will be assessed in the following sections.

The proposed kinematic accuracy link is a link between the accuracy of the trunk repositioning test and the accuracy of the position and motion parameters of the trunk during the trunk kinematics test. Table 20 shows the results of this analysis. There were both positive and negative correlations between the kinematic parameters and the AE in the different planes. There were two correlations that were statistically significant and at least 0.50 in strength. The correlation between the AE for axial rotation and the peak flexion accuracy during the kinematics test was –0.544 (p=0.036). This states that as the rotation accuracy in the axial rotation trunk repositioning test worsens with fatigue (i.e., the AE decreases) the average lumbar flexion angle during box liftoff decreases. The strongest correlation in this test was a correlation of –0.628 between the AE for flexion and the peak angular velocity in the coronal plane during the kinematics test. This states that as the flexion accuracy worsens with fatigue (i.e., the AE decreases) the peak lumbar lateral bending angle during box liftoff decreases.

Table 20: Correlation matrix to evaluate the kinematic accuracy link. Correlations significant at the p<0.05 level and with a minimum absolute value of 0.50 are listed in boldface type and marked with * and the strongest correlation is marked in italics.**

	Trunk Repositioning Flexion Accuracy	Trunk Repositioning Lateral Bending Accuracy	Trunk Repositioning Axial Rotation Accuracy
Lumbar Flexion	-0.379	0.066	**-0.544*** **
Lumbar Lateral Bending	-0.352	-0.309	0.201
Lumbar Axial Rotation	0.259	-0.327	-0.343
Lumbar Sagittal Angular Velocity	-0.489	-0.228	-0.430
Lumbar Coronal Angular Velocity	***-0.628**** **	-0.313	-0.154
Lumbar Transverse Angular Velocity	-0.307	0.122	0.017
Knee Flexion Angle	0.151	0.398	0.022

Table 21 shows the results from the kinematic precision link assessment. This link relates the precision of the trunk repositioning test with the precision of the position and motion parameters of the lumbar spine during the trunk kinematics test. The data exhibited a total of six statistically significant correlations stronger than 0.50 for the kinematic precision link assessment. The variability of the trunk repositioning test in each plane exhibited a significant correlation with at least one of the kinematic measures from the trunk kinematics test. The precision of the lateral bending trunk repositioning showed positive correlations with the variability of lumbar flexion during box liftoff (0.818), the variability of average knee flexion during box liftoff (0.856), and the variability of peak angular velocity in the transverse plane during box lifting (0.595). The precision of trunk repositioning in the other two planes showed negative correlations with the variability of the lumbar kinematic parameters. This suggests that as the trunk repositioning precision decreases (i.e., the variability increases) the variability of the kinematic measures decreases.

Table 21: Correlation matrix to evaluate the kinematic precision link. Correlations significant at the p<0.05 level and with a minimum absolute value of 0.50 are listed in boldface type and marked with * and the strongest correlation is marked in italics.**

	Trunk Repositioning Flexion Precision	Trunk Repositioning Lateral Bending Precision	Trunk Repositioning Axial Rotation Precision
Lumbar Flexion Precision	-0.049	**0.818*** **	**-0.549*** **
Lumbar Lateral Bending Precision	0.266	-0.202	0.005
Lumbar Axial Rotation Precision	-0.232	-0.205	-0.338
Lumbar Sagittal Angular Velocity Precision	**-0.568*** **	0.257	-0.399
Lumbar Coronal Angular Velocity Precision	**-0.551*** **	-0.167	0.192
Lumbar Transverse Angular Velocity Precision	-0.097	**0.595*** **	-0.229
Knee Flexion Angle Precision	-0.072	***0.856****	0.059

Table 22 shows the correlation coefficients for the kinetic accuracy link in the proposed model. This link suggests a relationship between the trunk force accuracy and the average NEMG measures. No statistically significant correlation was found between the force repetition accuracy and the various average NEMG measures. The highest correlation in this test was only 0.273, between the average NEMG for the trunk extensors during the sagittally symmetric muscle activity test and the force replication accuracy. This suggests that these values are fairly independent of one another in terms of how they responded to the fatiguing trunk extension exercise.

Table 22: Correlation matrix to evaluate the kinetic accuracy link. Correlations significant at the p<0.05 level and with a minimum absolute value of 0.50 are listed in boldface type and marked with * and the strongest correlation is marked in italics.**

	Flexor NEMG, symmetric test	**Extensor NEMG, symmetric test**	**Flexor NEMG, asymmetric test**	**Extensor NEMG, asymmetric test**
Force Replication Accuracy	0.070	*0.273*	0.104	-0.017

The last link in the proposed model is the kinetic precision link. This link suggests a relationship between the force replication precision and the variability of the NEMG measures. Table 23 shows that there was neither a correlation stronger than 0.50 nor a statistically significant correlation in this test. However, contradictory to the proposed model, there was a negative correlation between the variability of the averaged extensor NEMG in the asymmetric muscle activity test and the precision of the force replication test ($r = -0.492$, $p = 0.063$).

Table 23: Correlation matrix to evaluate the kinetic precision link. Correlations significant at the p<0.05 level and with a minimum absolute value of 0.50 are listed in boldface type and marked with * and the strongest correlation is marked in italics.**

	Flexor NEMG Precision, symmetric test	Extensor NEMG Precision, symmetric test	Flexor NEMG Precision, asymmetric test	Extensor NEMG Precision, asymmetric test
Force Replication Precision	0.085	0.120	0.161	*-0.492*

In addition to the correlations performed to test the links in the proposed models, two more correlation analyses were performed. These additional tests included the maximal trunk extension torque and reaction time data, neither of which were included in the original proposed model. Table 24 and Table 25 show this data.

Table 24 shows a highly significant strong negative correlation between the average lumbar lateral bending angle at box liftoff and the maximal trunk extension torque (r = -0.727, p=0.002). This states that as the trunk strength decreases, the value of lateral bending during liftoff increases. However, as previously mentioned, lateral bending towards the direction of the box is negative. Therefore, an increase in lateral bending value (i.e., becoming less negative) actually reduces the amount of lateral bending from neutral at box liftoff. For more information on the lateral bending angle, see the discussion on the effect of PERIOD on lateral bending angle in the results.

The average muscle onset time exhibited a number of significant correlations with average NEMG values from the muscle activity tests and with the average kinematics measures from the kinematics test. Table 22 shows significant positive correlations of the mean muscle onset time with the following kinematics measures: lateral bending angle during box liftoff (r = 0.553, p=0.032), rotation angle during box liftoff (r = 0.582, p=0.023), and peak lumbar sagittal angular velocity (r = 0.564, p=0.029) during the kinematics test.

Table 25 shows the correlations between the variability of the muscle onset times and the variability of the various NEMG and kinematics measures. There were no statistically significant correlations in this analysis. The strongest correlation was r=0.458 between the precision of the muscle onset time and the precision of the peak lumbar angular velocity in the coronal plane during the kinematics test (p = 0.086).

Figure 42 shows the number of significant correlations found in each link of the model diagram. This clearly shows that data more strongly supports the kinematic links as opposed to the kinetic links of the model.

Table 24: Correlation matrix with torque and mean muscle onset time. Correlations significant at the p<0.05 level and with a minimum absolute value of 0.50 are listed in boldface type and marked with * and the strongest correlation is marked in italics.**

	Flexor NEMG, symmetric test	Extensor NEMG, symmetric test	Flexor NEMG, asymmetric test	Extensor NEMG, asymmetric test	Lumbar Flexion	Lumbar Lateral Bending	Lumbar Axial Rotation
Maximal Trunk Torque	0.001	-0.034	0.088	-0.129	0.422	***-0.727****	0.069
Muscle Onset Time	**-0.532*****	0.427	-0.466	0.499	0.125	**0.553*****	**0.582*****

	Lumbar Coronal Angular Velocity	Lumbar Transverse Angular Velocity	Lumbar Sagittal Angular Velocity	Knee Flexion Angle
Maximal Trunk Torque	-0.349	-0.354	-0.024	0.267
Muscle Onset Time	0.492	0.471	**0.564*****	-0.197

Table 25: Correlation matrix with torque and muscle onset time variability. Correlations significant at the p<0.05 level and with a minimum absolute value of 0.50 are listed in boldface type and marked with * and the strongest correlation is marked in italics.**

	Flexor NEMG Precision, symmetric test	Extensor NEMG Precision, symmetric test	Flexor NEMG Precision, asymmetric test	Extensor NEMG Precision, asymmetric test	Lumbar Flexion Precision	Lumbar Lateral Bending Precision	Lumbar Axial Rotation Precision
Reaction Time Precision	-0.283	-0.367	-0.081	0.010	-0.0002	-0.185	0.176

	Lumbar Coronal Angular Velocity Precision	Lumbar Transverse Angular Velocity Precision	Lumbar Sagittal Angular Velocity Precision	Knee Flexion Angle Precision
Reaction Time Precision	*0.458*	0.197	0.000	-0.313

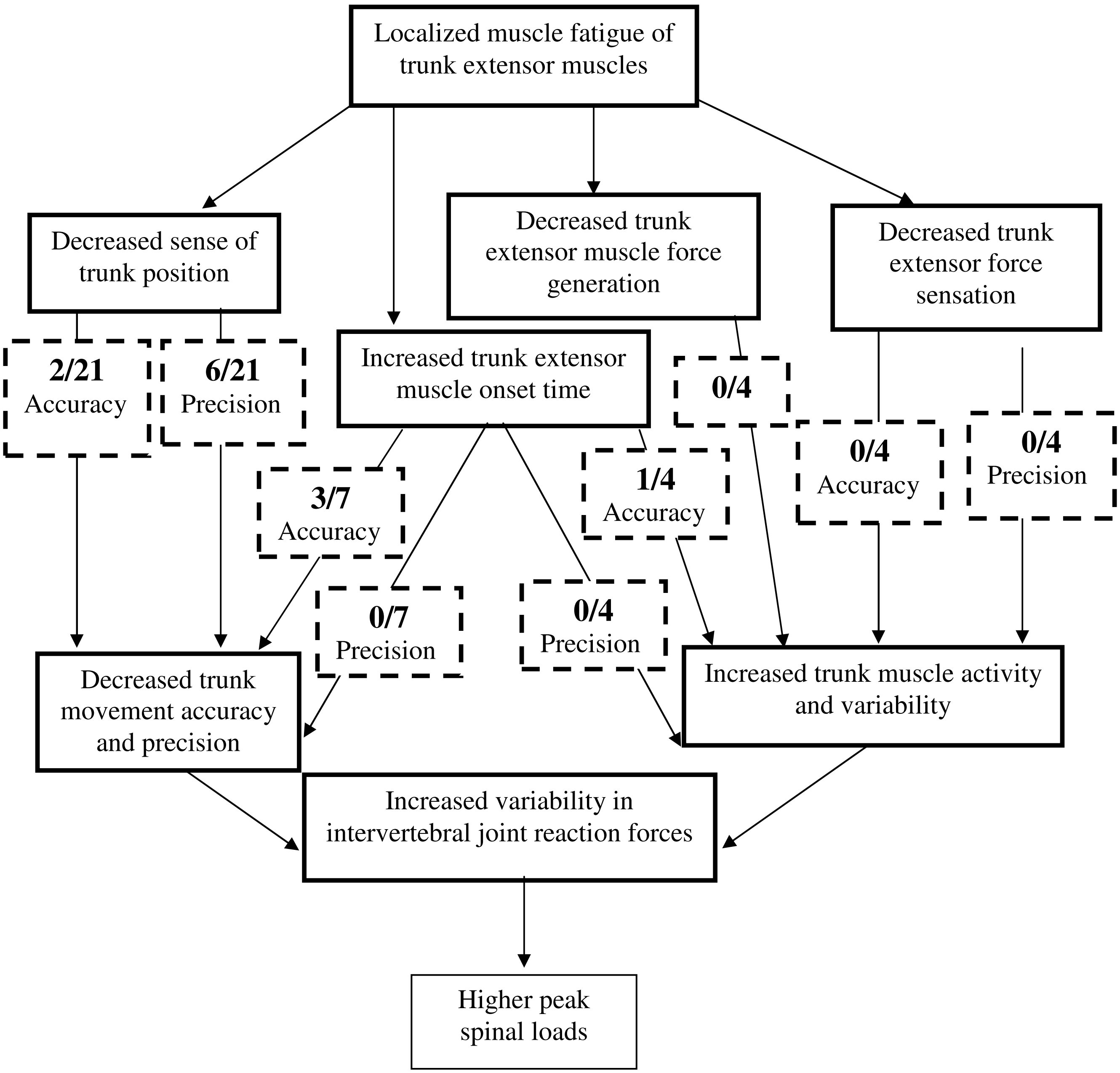

Figure 42: The proposed model of the effects of repetitive trunk extensions with the number of significant correlation coefficients for each link. The smaller number in each pair is the number of significant correlations and the larger number is the total number of correlations related to this link.

DISCUSSION

Verification of Fatigue

Several indices used in this experiment validated the assumption that the subjects did experience localized muscle fatigue of the trunk extensor muscles as a function of the experimental protocol. The Borg ratings of perceived exertion and the reduction in trunk torque during the maximal trunk extension trials showed that the experimental protocol did induce localized muscle fatigue. Borg CR-10 ratings have been shown to correlate well with other established objective measures of fatigue such as the median frequency of the raw EMG signal and the endurance time (Dedering et al, 1999). Conversely, the control subjects, whose main purpose was to ensure that just the time spent in the experiment was not the source of any changes in these dependent variables, did not show any changes with time into the experiment.

One important difference between the data in this experiment and data from previous research is that the subjects in this experiment did not progress all the way to exhaustion (e.g., Parnianpour et al, 1988). Since there were several tests being performed in the same experiment, it was important to ensure that the rate of fatigue generation was slow and consistent for all the different tests. Many other studies on low back muscle fatigue perform endurance studies where subjects are brought to a much further level of fatigue or even full failure. Dieen et al (1993a, 1993b) have used 25%, 40%, and 60% trunk extension MVC loads in different projects until subjects could no longer hold this load. Other researchers use the Biering-Sorenson trunk holding test wherein subjects must support their trunk horizontal using their trunk extensor muscles while their legs are strapped onto a table (Mannion et al, 1998). Other researchers have performed tests asking subjects to proceed to failure such that they could no longer perform the exercise at the intended load level (Parnianpour et al, 1988; Ng et al, 1997; Sparto et al, 1997b). Lastly, it is also important to note that there is a significant body of closely related literature that has used repeated lifting as the fatiguing task (Kim and Chung, 1995; Dieen et al, 1998; Sparto et al, 1997c; Dolan and Adams, 1998). Though very similar, the methodology using lifting as the fatiguing task allows the subjects to obtain less constrained body postures during the fatiguing task. This allows the subjects themselves to alter the progression and manifestations of fatigue in the experiment with different lifting styles. It is important to understand the differences in fatigue levels in previous research to understand how this research relates to the existing body of literature.

Trunk Repositioning Test

The trunk repositioning accuracy in the three planes all exhibited a significant effect of PERIOD in this experiment but the trunk repositioning precision data did not. The AE data found in this experiment during Period 1 is generally higher than that found in the literature for experiments without fatigue. The period 1 AE values seen in this study (7.0^o for flexion, 5.2^o lateral bending, 3.5^o for axial rotation) are higher than those found by Gill and Callaghan (1998), Swinkels and Dolan (2000), Koumantakis et al (2002), and Iwasa et al (2005). Only minor experimental procedural differences existed between this study and the ones just cited. However, the trend of larger AE in flexion than in lateral bending and in rotation found in this study does agree with the data of McGlashen et al (1991), Swinkels and Dolan (2000), Koumantakis et al (2002), and Iwasa et al (2005).

The finding of higher trunk repositioning absolute error after trunk extensor fatigue is in agreement with Iwasa et al (2005). They found AE to be higher in both lateral bending and forward flexion after fatigue (33% increase in group mean for flexion and 63% increase in lateral bending). This finding also agrees with the research of Tameila et al (1999). They found that the ability to sense a passive axial rotation in the lumbar spine was reduced after a set of fatiguing trunk extension exertions. Finally, this finding agrees with the larger body of research that shows decreases in proprioceptive capability with the onset of localized muscle fatigue (Skinner et al, 1986, Carpenter et al, 1998, Pedersen et al, 1999).

The underlying explanation for the decreased position sensation may be attributed to the decreased pH found in the muscles due to the buildup of lactic acid. Fukami (1988) and Fischer and Schafer (2005) have showed decreased muscle spindle activation when the spindles were bathed in fluids with decreased pH levels. Lactic acid buildup in the extracellular space can reduce the pH of these fluids, thus reducing the activation of the muscle spindles within these muscles.

Trunk Force Replication Test

The trunk force replication accuracy seems to be impacted by fatigue since there was a significant effect of PERIOD in the ANOVA results. This is especially interesting in light of some logistical challenges with a force perception test on the trunk. Most similar force perception research has been performed on the upper limb using a contralateral limb force-matching paradigm (e.g., Jones and Hunter, 1983a). In this type of test, the subject tries to match the force in one limb with the force in the other limb. One of the limbs is subjected to fatigue or some other treatment and the difference in force sensation between the two limbs can be attributed to the experimental treatment. However, this is obviously impossible to do with the trunk. Also, there is no viable way to remove the potential proprioceptive inputs from skin mechanoreceptors. The free nerve endings that lie near the surface of the skin can sense the amount of force on the back from the arm of the ARF during the testing to help the subject sense his force output. It is feasible that the underlying proprioceptive decrement may be even larger than seen in this experiment, and may be limited by the compensation of the free nerve endings in the back.

No other research has been identified that has used this test methodology, especially on the trunk, so it is impossible to relate these specific findings to previous research. However, the finding of a reduction in force sensation after fatiguing exertions does agree in general with the findings of Jones and Hunter (1983a, 1983b) and Jones (1986). This agreement with general findings using this novel test methodology suggests that this test methodology may be useful in further research on force perception, especially on the trunk where matched-force tests are impossible to perform.

Muscle Activity Test

Due to the similarity in the symmetric and asymmetric test modules, the discussion on the results for both of these modules will be combined here into one section. The increase in extensor NEMG in both the symmetric and asymmetric muscle activity tests supports existing research findings regarding the increase in NEMG with fatigue (Currier, 1969). The higher NEMG values in the extensors on the right side in the symmetric test suggest that the asymmetric testing and the asymmetric box lifting in the lifting kinematics test may have been fatiguing the right multifidus and right longissimus more than the muscles on the left side. However, since no real-time objective assessment of muscle fatigue (such as EMG median frequency calculations) was taken during the actual fatiguing trunk extension exertions there is no way to formally test this hypothesis.

It is interesting to note that there was not one flexor muscle that exhibited a statistically significant PERIOD effect on the NEMG values for either the symmetric or the asymmetric tests, although several of the oblique muscles neared statistical significance. Previous research has shown that antagonist muscles increase activity with agonist activity (Psek and Cafarelli 1993; Kelaher, 1996; Kellis 1999). It is possible that the lower level of fatigue that this experiment induced on these subjects may be the reason for the lack of stronger antagonist increases as the experiment progressed. This is an important differentiation between this study and other studies. Agonist muscle fatigue has been shown in previous research to heighten the sensitivity of the golgi tendon organs which, in turn, causes activation of the antagonist muscles via the interneurons arising from the Group 1b afferent motoneurons (McComas, 1996). The implication of this difference is in the control of stability. Recent research has suggested that antagonist muscle activity is related to enhancing stability of the spine (Granata et al, 2004). It is hypothesized that golgi tendon organs play a protective role in this enhanced stability. However, this mechanism may only be related to stronger levels of fatigue than that generated in this study and that is seen in typical industrial work.

The variability of the NEMG values was only affected by PERIOD in the asymmetric tests. One muscle (right rectus abdominis) showed a significant effect of PERIOD while three of the four external oblique muscle sections neared significance. This finding of increased variability with time into the experiment nicely complements the previous research on variability by Mirka and Marras (1993). Mirka and Marras (1993) found that increasing the angular velocity or the load on the trunk increases the variability of the NEMG patterns. This research is showing that the inclusion of fatigue in a trunk extension task also can increase the muscle NEMG variability, especially in the higher risk asymmetric actions. Therefore, the variability of several different inputs into the biomechanics of the trunk are increasing with increasing task demands (load, movement speed, asymmetry, and fatigue).

Trunk Kinematics Test

In general, all of the angular position posture and velocity measures in this study were found to be significantly impacted by the time into the experiment, but the angular acceleration values were not. A curious finding is the reduction in both lumbar flexion angle and knee flexion angle. These two measures are the primary drivers of how low the subjects' hands get when lifting a box. If both lumbar flexion and knee flexion decrease, then some other body part must be compensating for these reductions to allow the subjects' hands to get low enough to grab the box. One possible explanation is that the elbows and/or shoulders may be performing this compensatory postural adjustment. However, since no measures of arm position or orientation was collected in this experiment, it is impossible to determine if this actually happened in these subjects.

The post-hoc Tukey's tests on the trunk angles all showed a significant decrease in lumbar angles after Period 1 while the increases in peak lumbar angular velocity values did not occur until after Period 2. These two findings together suggest that the subjects were reducing their range of motion during the lifting task, but their angular velocities spiked higher during the lift. One explanation for the increase in angular velocities is that the subjects may have performed the entire lift in a slightly shorter time in Period 3 than in Periods 1 and 2. For this reason, a post-hoc analysis was performed on the box lifting time, defined as the difference in time from when the box left its origin and when it was deposited at its destination on the right side of the subject. The results of this analysis are shown in Figure 43. The subjects in the present experiment were lifting at a self-selected pace as though this was their job so the lifting speed should not have been minimized or maximized at any time in the experiment. Based on this supplemental analysis, the small (5%) but statistically significant decrease in lifting times may partially explain the increase in peak angular velocities in this experiment. It is possible that altered lifting patterns may also have contributed to this increase in peak angular velocities, but this can not be tested with this data. This finding does not necessarily agree with previous finding by Sparto et al (1997c) who found a non-significant increase of 8% in the lifting time generated from repetitive lifting. However, the participants in that study were instructed to lift at a maximal lifting rate until exhaustion (mean lifting frequency was 39 lifts/minute), so the participants were already trying to minimize their lift time at the beginning of the experiment.

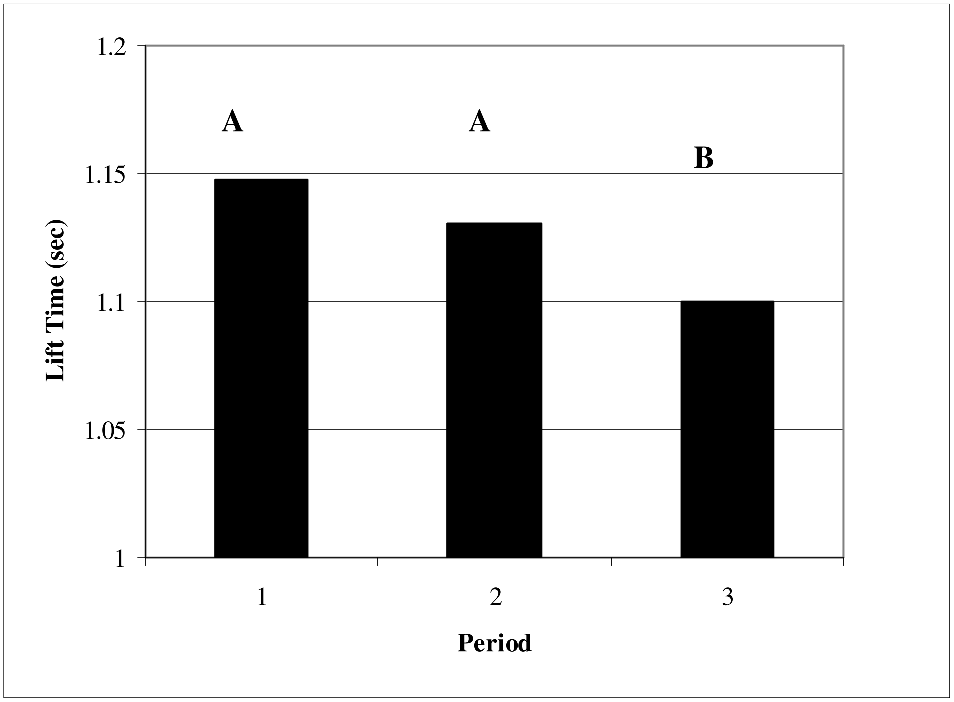

Figure 43: Box lifting time as a function of PERIOD for the experimental subjects.

The reduction in lift time does relate more closely to the work of Dolan and Adams (1998). Their experiment had subjects lifting a barbell in a sagittally-symmetric style at a self-selected pace. Their subjects' mean lifting frequency increased from 8.2 lifts/minute to 8.7 lifts/minute, which is much closer to the lifting rates used in this experiment.

The implications of these findings are that muscle fatigue impacted three of the five measures included in the low back injury risk model proposed by Marras et al (1993). Similarly, Marras et al (1999) have shown angular velocity and angular acceleration measures to more reliably distinguish between LBI patients and normals.

Muscle Onset Time Test

While only the left trunk extensor muscles exhibited a significant effect of PERIOD, all four extensor muscles trended towards increased muscle onset time as a function of test period for the experimental subjects. By contrast, the average extensor muscle onset time for the control subjects decreased as a function of PERIOD. It is likely that the control subjects were still learning and perfecting this task as the experiment progressed. Therefore, the experimental subjects may have had a greater increase in muscle onset time had the baseline been more stable. More practice on this specific task during the practice sessions may have prevented this from happening. However, the trends toward increasing muscle onset time with fatigue matches previous research. Wilder et al (1996) found a significant increase in muscle onset time from 87 msec to 108 msec with low back muscle fatigue. These data represent quicker muscle onset times and lower increases in muscle onset times as compared to the present study. Although the fatigue generated in their experiment was caused by vertical whole-body vibration during sitting and their load was applied by dropping a weight into a pan held in front of the abdomen, their trends were similar to that of the present study.

The cause of the increase in muscle onset time due to fatigue has been shown by Balestra et al (1992) to be related to the reduction or elimination of the short latency and medium latency components of the muscle reflex in the first dorsal interosseus. They hypothesized that inhibition of the α-motoneuron excitability. The baseline EMG values in the present research were had too high to identify the spikes in the EMG signal related to the specific latency components of the muscle reflex to visually validate this mechanism. More research where the subject is resting and his muscle activity was more quiet than in the present research is needed to visually verify this hypothesis.

Model Assessment

The theoretical model of interaction between biomechanics and proprioception during localized muscle fatigue was tested for a number of relationships among these interacting factors. The discussion that will follow will discuss the results seen in the experimental modules as they relate to this theoretical model.

The kinematic accuracy link showed two strong negative correlations. These two relationships involve motion in all three planes and involve both lifting posture at box liftoff and peak angular velocity. As the AE of trunk repositioning in the axial rotation increases with fatigue, the lumbar flexion at the point of box liftoff decreases. It is conceivable, though impossible to validate with this data, that the more laterally located muscles on the left side such as the longissimus, iliocostalis, are acting to stiffen the trunk which may reduce lumbar flexion may also be contributing to the reduced proprioceptive capability of the extensor muscles. This is supported by the fact there is also a negative correlation, though weaker in strength at r=-0.379, between the trunk repositioning AE in flexion and the average lumbar flexion at box liftoff in the kinematics test.

A stronger negative relationship was found between the peak lumbar angular velocity in the coronal plane and the AE of trunk repositioning in trunk flexion. As the mean of the peak lumbar angular velocity in the coronal plane increased, the trunk repositioning AE in forward flexion decreased. This is difficult to explain as both measures increased with time into the experiment which suggests they react similarly to PERIOD yet their correlation was negative, suggesting they react differently. It is interesting to note that none of the angle positions at box liftoff had significant correlations with the trunk repositioning errors. One might expect a closer relationship between these variables as they are both in the angle domain.

The results in the precision kinematic link show more significant correlations than the accuracy kinematic link. However, there is a fair degree of similarity between the two correlation matrices in terms of the sign and relative magnitude of the correlations. This suggests that many of the accuracy and precision measures may be acting similarly as a function of fatigue. This is supported by the graphical analysis by comparing trunk repositioning accuracy in Figure 36 to the trunk repositioning precision in Figure 37 for flexion and rotation. The graphs in these two figures are very similar. The reason for this similarity may be attributed to the design of the experiment. It is feasible that the calibration repetitions performed at the beginning of each force replication or trunk repositioning test were preventing any long-term drift that might reduce the subject's accuracy, but not his precision. These calibration repetitions were felt to be necessary due to the number of tasks the subjects performed and the time between subsequent sets of force replication or trunk repositioning tests. No previous research was identified that compared the response of accuracy measures with the response of precision measures for a proprioceptive task. Future work to separate the difference between accuracy and precision without these calibration repetitions may be needed to further refine this model.

The precision of the trunk flexion repositioning was significantly correlated with the variability of the peak lumbar flexion in the sagittal and coronal planes. This is an important finding as the angular velocities in these two planes are two of the five risk factors identified by Marras et al (1993) in their field study of low back injury risk based on dynamic manual material handling jobs. Therefore, higher variability in one or both of these variables can increase the predicted low back injury risk as calculated by this risk model. Biomechanically, this also makes sense as the amount of co-contraction of the shear-generating muscles such as the oblique muscles will increase with higher angular velocities (Marras and Mirka, 1992) during the higher peak velocities.

The kinetic accuracy and kinetic precision links did not show any significant strong correlations among the variables. It is odd that there would be no high correlation with the force replication test accuracy since it was significantly affected by PERIOD. Without previous research using this experimental methodology, it is difficult to provide any insight into the validity of this methodology. However, the methods used to measure the muscle activity accuracy and precision are well tested.

While the precision of muscle onset time did not provide any significant correlates, the accuracy of the muscle onset time yielded 4 significant correlations out of a total of 11. Interestingly, the data support the construct that the muscle onset time relates both to the kinematic part of the model (3/7 significant correlations) and the kinetic part of the model (1/4 significant correlations). This supports the concept that the muscle onset time bridges the gap between the kinetic and kinematic measures of trunk biomechanics during fatigue.

The muscle onset time and sagittal angular velocity both showed fairly consistent linearity among the individual subjects (see Appendix G). This led to a significant correlation of 0.564 meaning as the peak of the angular velocity increases, the muscle onset time also increases. It is hypothesized that the fatigue in the fast-twitch muscle fibers of the trunk slowed the activation of these muscles in the muscle onset time. This reduction in muscle onset time also impacted the peak angular velocity by delaying the range of motion of the trunk during the lift. This relative delaying of the trunk extension to later in the lift has been shown to result from repetitive lifting by Sparto et al (1997c).

Overall, the data provided strong support for the kinematic links and the muscle onset time links in the model, but very little support for the kinetic links in the model. It is feasible that the correlations in the kinetic links might be strengthened with a more severe level of fatigue being induced in the subjects. However, it is felt that more severe level of fatigue may be less representative of the type of fatigue experienced by industrial workers.

The implications of these findings and the utility of this model will be discussed. The very simplistic model does not totally reflect the complex nature of the neurophysiology and biomechanics of the trunk system. However, the model was supported by the data for the kinematic portion of the model. It is hypothesized that with lower level fatigue of gradual onset, the postural and motion parameters may be of more concern than the reduction in strength and altered muscle patterns. These findings relate more closely to industrial workers who often perform manual jobs with some transient minor levels of fatigue throughout the day but never attain muscular exhaustion or total muscle fatigue. This suggests that more research needs to focus on this lower level and gradual onset fatigue to further understand the subtleties of the interactive effects on the individual risk of low back injury.

Limitations

As with any other research study, this experiment has a number of limitations that must be mentioned. One of the key limitations of this study was the low number of subjects. Due to the amount of testing involved for each subject, the number of subjects was kept small. Also, the inclusion of control subjects was key to ensure that any changes in the dependent variables was in fact due to muscle fatigue and not due to any other factor. However, the time needed to run the control subjects further limited the number of experimental subjects. Another limitation of this study was the use of a homogeneous subject pool. Since variability in the dependent variables was of interest in this study, it was important to have a homogeneous subject pool to limit differences due to body size, age, and fitness level. As a result of this small sample size, a fixed effects statistical model was used to evaluate the effect of PERIOD on the independent measures. This choice eliminates the possibility of extrapolating these results to the larger population. However, since this research is exploratory in nature, searching for possible interactive links between effects of fatigue, this choice seemed logical to help explain the data more clearly.

Another limitation inherent in this experiment is the lack of control over the progression of fatigue during the entire experiment. While a lot of pilot work went into the determination of the final test order, it is impossible to know the exact pattern of fatigue during the experiment since there is no single measure of fatigue that could have been collected continuously throughout the experiment. It is possible that some of the less stressful tests, such as the trunk repositioning test, allowed some transient recovery from fatigue. Similarly, it is possible that the amount of fatigue was increasing during the more stressful tests such as the muscle activity tests.

The manner in which the fatigue was generated can be considered a limitation of the study since only one method of fatigue could be performed whereas fatigue can be generated in an infinite number of ways. For example, fatigue could have been generated asymmetrically which would presumably have fatigued the contralateral side of the trunk extensors more quickly than the ipsilateral side relative to the starting trunk position. Symmetric fatigue was chosen since many jobs require asymmetric lifting from both sides and it was determined that symmetric fatigue would adequately simulate this type of fatigue. One logistical consideration with asymmetric fatigue is that each test result would require both contralateral and ipsilateral testing directions to understand the effects of the fatigue on each direction of motion or force. For example, this would have required trunk repositioning tests for both lateral bending and twisting to be performed in both directions. This extra testing was seen as negatively impacting the generation of fatigue during the experiment as a whole. Further focused research should look at the effects of asymmetric fatigue.

Another limitation of this study related to control was the use of controlled isokinetic trunk extensions to generate fatigue. While seemingly in conflict with the limitation noted before this one, the artificial nature of the trunk extensions in the ARF is not fully representative of real world work. However, this device and similar devices have been used in a number of studies in order to generate a steady level of fatigue on the subjects. Also, the inclusion of two training sessions and a practice block performed the day of the experiment before data collection mitigated any learning effect during the data collection session.

A further limitation on the study involved the logistics of the experiment. In an experiment where muscle fatigue is being generated, the time between the fatiguing task and the test should be minimized so that no significant recovery could occur between the fatiguing task and the evaluation test. However, the equipment needed for this experiment dictated that the subject enter and exit the ARF repeatedly throughout the experiment. The trunk repositioning test and the kinematics test could not be performed in the ARF due to the amount of metal in the ARF structure. This metal in the ARF would interfere with the signals from the MotionStar electromagnetic sensors so much as to render the position and motion data meaningless. However, a lot of pilot work looked at streamlining the logistics of running the experiment so as to minimize the time spent entering and exiting the ARF. Also, each subject was tested with the help of a research assistant to further help minimize this wasted time during the experiment.

Future Research

As mentioned in the Limitations section the small sample size and the subsequent choice of a fixed effects statistical model limit the ability to extrapolate this data to the population. Therefore, a replication of the existing study with a larger diverse subject population including females would be needed to be able to extrapolate these results to the population in general.

The findings of this study and previous research provide evidence for the importance of proprioception measures as a function of fatigue. It is felt that more research overall in the various areas of proprioception is needed to more fully understand the effects of manual materials handling work and localized muscle fatigue.

Based on the findings of this study, many different follow-up studies can be performed. The model proposed in this research only investigates correlations between different factors, not causation. Several individual follow-up investigations can be performed to isolate single relationships to more thoroughly understand whether there is a direct link between these factors or if there are other factors that intermediate between them.

Lastly, the link between muscle fatigue and the risk of injury needs to be investigated. A lot of research has been performed on the contributors to muscle fatigue and the effects of muscle fatigue. However, very little research has been performed that links muscle fatigue to risk of injury. If a strong link between muscle fatigue and risk of injury can be established, it would significantly increase our understanding of the risk of injury and our ability to test hypotheses related to increased or reduced injury risk.

CONCLUSION

This study used a holistic approach to investigate the effects of localized muscle fatigue of the trunk extensors of the back. The interaction between the biomechanical and proprioceptive effects of localized muscle fatigue have not been previously studied in a single project. This research included tests of both proprioceptive and biomechanical nature, providing a more complete assessment of the effects of localized muscle fatigue. While the number of subjects in this study was small, the individual results in this study agree with previous findings and extend them further. The data showed that localized muscle fatigue of the trunk reduced trunk positioning accuracy, lifting kinematics accuracy, muscle onset time, force replication accuracy, and increased trunk extensor muscle activity. In order to describe the inter-relationship between the various biomechanical and proprioceptive effects of localized muscle fatigue, a theoretical model was proposed. The data in the study supports portions of the model primarily related to kinematics and muscle onset time while the kinetic data did not support the relationships proposed in the theoretical model. It is suggested that the kinematic effects of localized muscle fatigue may appear before the kinetic effects appear, especially in minor localized fatigue that does not lead to total exhaustion. The implications to industry of these findings are that there are a variety of responses to localized muscle fatigue that ergonomists and industrial engineers must understand while designing jobs and work methods. Also, this research suggests that the earlier manifestations of localized muscle fatigue of the low back may not be strength and muscular in nature, but more related to posture and motion.

REFERENCES

Ascension Technology Corporation (2000) MotionStar ® web page, technical specifications: http://www.ascension-tech.com/graphic.htm.

Adams, M.A. and Hutton, W.C. (1982) Prolapsed intervertebral disc: A hyperflexion injury. *Spine*, 7(3):184-191.

Adams, M.A. and Hutton, W.C. (1983) The effect of posture on the fluid content of lumbar intervertebral discs. *Spine*, 8(6):665-671.

Allison, G.T. and Fukushima, S. (2003) Estimating three-dimensional spinal repositioning error: The impact of range, posture, and number of trials. *Spine*, 28(22):2510-2516.

Andersson, G.B.J., Pope, M.H., and Frymoyer, J.W. (1984) Epidemiology. In *Occupational Low Back Disorders* (Edited by Andersson, Pope, and Frymoyer), Praeger Press, 101-113.

Balestra, C., Duchateau, J., and Hainut, K. (1992) Effects of fatigue on the stretch reflex in a human muscle. *Electroencephalography and Clinical Neurophysiology*, 85:46-52.

Basmajian, J.V. and DeLuca, C.J. (1985) *Muscles Alive: Their Functions Revealed by Electromyography*. Baltimore, MD: Williams and Wilkins.

Battie, M.C., Videman, T., Gill, K., Moneta, G.B., Nyman, R., Kaprio, J., and Koskenvuo, M. (1991) Smoking and lumbar intervertebral disc degeneration: An MRI study of identical twins. *Spine*, 16(9):1015-1021.

Biering-Sorensen, F., Thomsen, C.E., and Hilden, J. (1989) Risk indicators for low back trouble. *Scandinavian Journal for Rehabilitative Medicine*, 21:151-157.

Bigland, B, and Lippold, O.C.J. (1956) The relation between force velocity and integrated electrical activity in human muscles. *Journal of Physiology*, 123:214-224.

Bogduk, N. (1983) The innervation of the lumbar spine. Spine, 8(3):286-293.

Borg, G. (1998) *Borg's Perceived Exertion and Pain Scales.* Stockholm: Human Kinetics.

Bureau of Labor Statistics (2002) Lost-worktime injuries and illnesses: Characteristics and resulting time away from work, 2000. Washington, D.C.: United States Department of Labor.

Carpenter, J.E., Blasier, R.B., and Pellizzon, G.G. (1998) The effects of muscle fatigue on shoulder joint position sense. *American Journal of Sports Medicine*, 26(2):262-265.

Chaffin, D.B. and Park, K.S. (1973) A longitudinal study of low-back pain as associated with occupational weight lifting factors. *American Industrial Hygiene Association Journal*, 34(12):513-525.

Cholewicki, J., Greene, H.S., Polzhofer, G.K., Galloway, M.T., Shah, R.A., and Radebold, A. (2002) Neuromuscular function in athletes following recovery from a recent acute low back injury. *Journal of Orthopaedic and Sports Physical Therapy*, 32(11):568-575.

Daltroy, L.H., Larson, M.G., Wright, E.A., Malspeis, S., Fossel, A.H., Ryan, J., Zwerling, C., and Liang, M.H. (1991) A case-control study of risk factors for industrial low back injury: Implications for primary and secondary prevention programs. *American Journal of Industrial Medicine*, 20:505-515.

Descarreaux, M., Blouin, J.S., and Teasdale, N. (2004) Force production parameters in patients with low back pain and healthy control study participants. *Spine*, 29(3):311-317.

Dedering, A., Nemeth, G., and Harms-Ringdahl, K. (1999) Correlation between electromyographic spectral changes and subjective assessment of lumbar muscle fatigue in subjects without pain from the lower back. *Clinical Biomechanics*, 14:103-111.

Dedering, A., Hjelmsater, M., Elfving, B., Harms-Ringdahl, K., and Nemeth, G. (2000) Between-days reliability of subjective and objective assessments of back extensor muscle fatigue in subjects without lower-back pain. *Journal of Electromyography and Kinesiology*, 10:151-158.

Deyo, R.A. and Bass, J.E. (1989) Lifestyle and low-back pain. *Spine*, 14(5):501-506.

Dieen, J.H. van, Vrielink, H.H.E.O., and Toussaint, H.M. (1993a) An investigation into the relevance of the pattern of temporal activation with respect to erector spinae muscle endurance. *European Journal of Applied Physiology and Occupational Physiology*, 66:70-75.

Dieen, J.H. van, Vrielink, H.H.E.O., Housher, A.F., Lotters, F.B.J., and Toussaint, H.M. (1993b) Trunk extensor endurance and its relationship to electromyogram parameters. *European Journal of Applied Physiology and Occupational Physiology*, 66:388-396.

Dieen, J.H. van, (1996) Asymmetry of erector spinae muscle activity in twisted postures and consistency of muscle activation patterns across subjects. *Spine*, 21(22):2651-2661.

Dieen, J.H. van, Boke, B., Oosterhuis, W., and Toussaint, H.M. (1996) The influence of torque and velocity on erector spinae muscle fatigue and its relationship to changes of electromyogram spectrum density. *European Journal of Applied Physiology and Occupational Physiology*, 72(4):310-315.

Dieen, J.H. van, Burg, P. van der, Raaijmakers, T.A.J., and Toussaint, H.M. (1998) Effects of repetitive lifting on kinematics: Inadequate anticipatory control or adaptive changes. *Journal of Motor Behavior*, 30(1):20-32.

DiFabio, R.P. (1987) Reliability of computerized surface electromyography for determining the onset of muscle activity. *Physical Therapy*, 67(1):43-48.

Dolan, P. and Adams, M.A. (1998) Repetitive lifting tasks fatigue the back muscles and increase the bending moment acting on the lumbar spine. *Journal of Biomechanics*, 31:713-721.

Ebig, M., Lephart, S.M., Burdett, R.G., Miller, M.C., and Pinciverio, D.M. (1997) The effect of sudden inversion stress on EMG activity of the peroneal and tibialis anterior muscles in the chronically unstable ankle. *Journal of Orthopaedic and Sports Physical Therapy*, 26(20:73-77.

Fathallah, F.A., Marras, W.S., and Parnianpour, M. (1998a) An assessment of complex spinal loads during dynamic lifting tasks. *Spine*, 23(6):706-716.

Fathallah F.A., Marras W.S., Parnianpour M. (1998b) The role of complex, simultaneous trunk motions in the risk of occupation-related low back disorders. *Spine*, 23(9):1035-1042.

Fischer, M. and Schafer, S.S. (2005). Effects of changes in pH on the afferent impulse activity of isolated cat muscle spindles. *Brain Research*, 1043(1):163-178.

Flor, H., Schugens, M.M., and Birbauer, N. (1992) Discrimination of muscle tension in chronic pain patients and healthy controls. *Biofeedback and Self-Regulation*, 17(3):165-177.

Fukami, Y. The effects of NH3 and CO2 on the sensory ending of mammalian muscle spindles: intracellular pH as a possible mechanism. *Brain Research*, 463(1):140-143.

Garg, A., and Saxena, U. (1979) Effects of lifting frequency and technique on physical fatigue with special reference to psychophysical methodology and metabolic rate. *American Industrial Hygiene Association Journal*, 40:894-903.

Gill, K.P. and Callaghan, M.J. (1998) The measurement of lumbar proprioception in individuals with and without low back pain. *Spine*, 23(3):371-377.

Glasscock, N.F. (2003) *Exploring the relationships between psychosocial factors, biomechanical workstyle, muscle tension, and musculoskeletal discomfort reporting.* Unpublished doctoral dissertation, North Carolina State University, Raleigh, NC

Granata, K.P., Slota, G.P., and Wilson, S.E. (2004) Influence of fatigue in neuromuscular control of spinal stability. *Human Factors*, 46(1):81-91.

Hagen, K.B., Sorhagen, O., and Harms-Ringdahl, K. (1995) Influence of weight and frequency on thigh and lower-trunk motion during repetitive lifting techniques employing stoop and squat techniques. *Clinical Biomechanics*, 10(3):122-127.

Han, T.S., Schouten, J.S.A.G., Lean, M.E.J., and Seidell, J.C. (1997) The prevalence of low back pain and associations with body fatness, fat distribution and height. *International Journal of Obesity*, 21:600-607.

Herrin, G.D., Hariedi, M., and Anderson, C.K. (1986) Prediction of overexertion injuries using biomechanical and psychophysical models. *American Industrial Hygiene Association Journal*, 47:322-330.

Hides, J.A., Stokes, M.J., Saide, M., Jull, G.A., and Cooper, D.H. (1994) Evidence of lumbar multifidus muscle wasting ipsilateral to symptoms in patients with acute/subacute low back pain. *Spine*, 19(2):165-172.

Iwasa, K., Miyamoto, K., and Shimizu, K. (2005) Effects of erector spinae muscle fatigue on trunk repositioning accuracy in forward and lateral flexion. *Journal of Back and Musculoskeletal Rehabilitation*, 18:61-66.

Jones, L.A. and Hunter, I.W. (1983a) Effect of fatigue on force sensation. *Experimental Neurology*, 81:640-650.

Jones, L.A. and Hunter, I.W. (1983b) Perceived force in fatiguing isometric contractions. *Perception and Psychophysics*, 33(4):369-374.

Jones, L.A. (1986) Perception of force and weight: Theory and research. *Psychological Bulletin*, 199(1):29-42.

Kelaher, DP. (1996) The effects of training history and muscle fatigue on load sharing in a multiple muscle system, *Proceedings of the Human Factors and Ergonomics Society 40th Annual Meeting*, 589-593.

Kelaher, D.P., Glasscock, N.F., and Mirka, G.A. (2003) The effects of body awareness, movement direction, visual condition, and movement speed on trunk proprioception, *Proceedings of the Human Factors and Ergonomics Society 47th Annual Meeting*, 610-614.

Keller, T.S., Holm, S.H., Hansson, T.H., and Spengler, D.M. (1990) The dependence of intervertebral disc mechanical properties on physiologic conditions. *Spine*, 15(8):751-761.

Kellis E (1999) The effects of fatigue on the resultant joint moment, agonist and antagonist electromyographic activity at different angles during dynamic knee extension efforts. *Journal of Electromyography and Kinesiology* 9:191-199.

Kelsey, J.L. Githens, P.B., Walter, S.D., Southwick, W.O., Weil, U., Holford, T.R., Ostfeld, A.M., Calogero, J.A., O'Connor, T., White, A.A. (1984) An epidemiological study of the relationship between occupations and acute herniated lumbar intervertebral discs. *Journal of Bone and Joint Surgery*, 66(6):907-914.

Kim, S.H. and Chung, M.K. (1995) Effects of posture, weight, and frequency on trunk muscular activity and fatigue during repetitive lifting tasks. *Ergonomics*, 38(5):853-863.

Koumantakis , G.A., Winstanley, J., and Oldham, J.A. (2002) Thoracolumber proprioception in individuals with and without low back pain: Intratester reliability, clinical applicability, and validity. *Journal of Orthopedic and Sports Physical Therapy*, 32(7):327-335.

Lavender, S.A. Mirka, G.A., Schoenmarklin, R.W., Sommerich, C.M., Sudhakar, L.R., and Marras, W.S. (1989) The effects of preview and task symmetry on trunk muscle response to sudden loading. *Human Factors*, 31(1):101-115.

Lavender, S.A., Tsuang, Y.H., Andersson, G.B.J., Hafezi, A., and Shin, C.C. (1992) Trunk muscle cocontraction: The effects of moment direction and moment magnitude. *Journal of Orthopaedic Research*, 10(5):691-700.

Lavender, S.A., Marras, W.S., and Miller, R.A. (1993) The development of response strategies in preparation for sudden loading to the torso. *Spine*, 18(14):2097-2105.

Lavender, S.A. and Marras, W.S. (1994) The use of turnover rate as a passive surveillance indicator for potential low back disorders. *Ergonomics*, 37(6):971-978.

Lawrence, B.M., Mirka, G.A., and Buckner, G.D. (2005) Adaptive system identification applied to the biomechanical response of the human trunk during sudden loading. *Journal of Biomechanics*, 38:2472-2479.

Lippold, O.C.J. (1952) The relation between integrated action potentials in the human muscle and its isometric tension. *Journal of Physiology*, 117:492-499.

Magnusson, M.L., Aleksiev, A.R., Wilder, D.G., Pope, M.H., Spratt, K.F., Lee, S.H., Goel, V.K., and Weinstein, J.N. (1997) Unexpected load and asymmetric posture as etiologic factors in low back pain. *European Spine Journal*, 5(1):23-35.

Mair, S.D., Seaber, A.V., Glisson, R.R., and Garret, W.E. (1996) The role of fatigue in susceptibility to acute muscle strain injury. *American Journal of Sports Medicine*, 24(2):137-143.

Manning, D.P., Mitchell, R.G., and Blanchfield, L.P. (1984) Body movements and events contributing to accidental and nonaccidental back injuries. *Spine*, 9(7):734-739.

Mannion, A.F., Adams, M.A., and Dolan, P. (2000) Sudden and unexpected loading generates high forces on the lumbar spine. *Spine*, 25(7):842-852.

Mannion, A.F., Dumas, G.A., Stevenson, J.M., and Cooper, R.G. (1998) The influence of muscle fiber size and type distribution on electromyographic measures of back muscle fatigability. *Spine*, 23(5):576-584.

Marras, W.S. (1998) Point of view. *Spine*, 23(7):781.

Marras, W.S. and Mirka, G.A. (1989) Trunk strength during asymmetric trunk motion, *Human Factors,* 31(6):667-677.

Marras, W.S. and Mirka, G.A. (1992) A comprehensive evaluation of asymmetric trunk motions. *Spine*, 17:318-326.

Marras, W.S., Rangarajulu, S.L., and Lavender, S.L. (1987) Trunk loading and expectation. *Ergonomics*, 30(3):551-562.

Marras, W.S., Lavender, S.A., Leurgans, S.E., Rajulu, S.L., Allread, W.G., Fathallah, F.A., and Ferguson, S.A. (1993) The role of dynamic three-dimensional trunk motion in occupationally-related low back disorders: The effects of workplace factors, trunk position, and trunk motion characteristics on risk of injury. *Spine*, 18(5):617-628.

Marras, W.S. and Granata, K.P. (1997) Changes in trunk dynamics and spine loading during repeated trunk exertions. *Spine*, 22(21):2564-2570.

Marras, W.S., Ferguson, S.A., Gupta, P., Bose, S., Parnianpour, M., Kim, J., and Crowell, R. (1999). The quantification of low back disorder using motion measures: Methodology and validation. *Spine*, 24(20):2091-2100.

McComas, A.J. (1996) *Skeletal Muscle: Form and Function.* Champaign, IL: Human Kinetics.

McGlashen, K., Ashton-Miller, J.A., Green, M., and Schultz, A.B. (1991) Trunk positioning accuracy in the frontal and sagittal planes. *Journal of Orthopedic Research*, 9(4):576-583.

McLain, R.F., and Pickar, J.G. (1998) Mechanoreceptor endings in human thoracic and lumbar facet joints. *Spine*, 23(2):168-173.

McNair, P.J. and Heine, P.J. (1999) Trunk proprioception: Enhancement through lumbar bracing. *Archives of Physical Medicine and Rehabilitation*, 80:96-99.

Mirka, G.A. (1991) The quantification of EMG normalization error. *Ergonomics*, 34(3), 343-352.

Mirka, G.A. and Baker, A. (1996) An investigation of the variability in human performance during sagittally symmetric lifting tasks. *IIE Transactions*, 28:745-752.

Mirka, G.A. and Marras, W.S. (1993) A stochastic model of trunk muscle coactivation during trunk bending. *Spine*, 18(11):1396-1409.

Mirka, G.A., Kelaher, D.P., Baker, A., Harrison, A.E., and Davis, J.R. (1997) Selective activation of the external oblique musculature during axial torque production. *Clinical Biomechanics*, 12(3):172-180.

Montgomery, D.C. (2005) *Design and Analysis of Experiments, 6th edition*, New York, NY: John Wiley and Sons.

Nerlich, Andreas G., Schliecher, Erwin D., and Boos, Norbert. (1997) Immuno-histologic markers for age-related changes of human lumbar intervertebral discs. *Spine*, 22(24):2781-2795.

Newcomer, K., Laskowski, E.R., Yu, B., Larson, D.R., and An, K.-N. (2000) Repositioning error in low back pain. Comparing trunk repositioning error in subjects with chronic low back pain and controls subjects. *Spine*, 25(2):245-250.

Ng, J.K., Richardson, C.A., and Jull, G.A. (1997) Electromyographic amplitude and frequency changes in the iliocostalis lumborum and multifidus muscles during a trunk holding test. *Physical Therapy*, 77(9):954-961.

Omino, K., and Hayashi, Y. (1992) Preparation of dynamic posture and occurrence of low back pain. *Ergonomics*, 35(5-6):693-707.

O'Brien, P.R., and Potvin, J.R. (1997) Fatigue-related EMG responses of trunk muscles to a prolonged, isometric twist exertion. *Clinical Biomechanics*, 12(5):306-313.

Parkhurst, T.M. and Burnett, C.N. (1994) Injury and proprioception in the lower back. *Journal of Orthopedic and Sports Physical Therapy*, 19(5):282-295.

Parnianpour, M., Nordin M., Kahanovitz N., and Frankel V. (1988) The triaxial coupling of torque generation of trunk muscles during isometric exertions and the effect of fatiguing isoinertial movements on the motor output and movement patterns. *Spine*, 13(9):982-992.

Peck, C.L. and Kraft, G.H. (1977) Electromyographic biofeedback for pain related to muscle tension. A study of tension headache, back, and jaw pain. *Archives of Surgery*, 112(7):889-895.

Pedersen, J., Lonn, J., Hellstrom, F., Djupsjobacka, M., and Johansson, H. (1999) Localized muscle fatigue decreases the acuity of the movement sense in the human shoulder. *Medicine and Science in Sports and Exercise*, 31(7):1047-1052.

Potvin, J.R. (1992) *The Influences of Fatigue on Hypothesized Mechanisms of Injury to the Low Back during Repetitive Lifting.* PhD Dissertation, Waterloo, Ontario: University of Waterloo.

Potvin, J.R. and O'Brien, P.R. (1998) Trunk muscle co-contraction increases during fatiguing, isometric, lateral bend exertions. Possible implications for spine stability. *Spine*, 23(7):774-780.

Psek, J.A., and Cafarelli, E. 1993. Behavior of coactive muscles during fatigue. *Journal of Applied Physiology,* 74(1):170-175.

Punnett, L., Fine, L.J., Keyserling, W.M., Herrin, G.D., and Chaffin, D.B. (1991) Back disorders and nonneutral trunk postures of automobile assembly workers. *Scandinavian Journal of Work and Environmental Health*, 17:337-346.

Radebold, A., Cholewicki, J., Panjabi, M.M., and Patel, T.C. (2000) Muscle response pattern to sudden trunk loading in healthy individual and in patients with chronic low back pain. *Spine*, 25(8)947-954.

Shields, S.A., Mallory, M.E., & Simon, A. (1989) The body awareness questionnaire: reliability and validity. *Journal of Personality Assessment, 53*(4), 802-815.

Skinner, H.B., Wyatt, M.P., Hodgon, J.A., Conard, D.W., and Barrack, R.L. (1986) Effect of fatigue on joint position sense of the knee. *Journal of Orthopaedic Research*, 4(1):112-118.

Smith-Jackson, T. (1998) *Irrelevant speech, verbal task performance, and focused attention: a laboratory examination of the performance dynamics of open-plan offices.* Unpublished doctoral dissertation, North Carolina State University, Raleigh, NC.Sparto, P.J., Parnianpour, M., Reinsel, T.E., and Simon, S. (1997a) Spectral and temporal responses of trunk extensor electromyography to an isometric endurance test. *Spine*, 22(4):418-425.

Sparto, P.J., Parnianpour, M., Marras, W.S., Granata, K.P., Reinsel, T.E., and Simon, S. (1997b) Neuromuscular trunk performance and spinal loading during a fatiguing isometric trunk extension with varying torque requirements. *Journal of Spinal Disorders*, 10(2):145-156.

Sparto, P.J., Parnianpour, M., Reinsel, T.E., and Simon, S. (1997c) The effect of fatigue on multijoint kinematics, coordination, and postural stability during a repetitive lifting task. *Journal of Orthopedic Sports Physical Therapy*, 25(1):3-12.

Sterner, R.L., Pincivero, D.M., Lephart, S.M. (1998) The effects of muscular fatigue on shoulder proprioception. *Clinical Journal of Sport Medicine*, 8:96-101.

Swet, J.E. and Schoutz, T.W. (1975) Mechanical transduction in the Golgi tendon organ. A hypothesis. *Archives Italiennes de Biologie*, 374-382.

Swinkels, A. and Dolan, P. (1998) Regional assessment of joint position sense in the spine. *Spine*, 23(5):590-597.

Swinkels, A. and Dolan, P. (2000) Spinal position sense is independent of the magnitude of movement. *Spine*, 25(1):98-105.

Taimela, S., Kankaanpaa, M., and Luoto, S. (1999) The effect of lumbar fatigue on the ability to sense a change in lumbar position. *Spine,* 24(13): 1322-1327.

Thelen, D.G., Schultz, A.B., and Ashton-Miller, J.A. (1995) Co-contraction of lumbar muscles during the development of time-varying triaxial moments. *Journal of Orthopaedic Research*, 13(3):390-398.

Troup, J.D., Martin, J.W., and Lloyd, D.C. (1981) Back pain in industry. A prospective survey. *Spine*, 6(1):61-69.

Waters, T.R., Putz-Anderson, V., Garg, A., and Fine, L.J. (1993) Revised NIOSH equation for the design and evaluation of manual lifting tasks. *Ergonomics*, 36(7):749-776.

Webster, B.S. and Snook, S.H. (1994) The cost of 1989 Workers' Compensation low back pain claims. *Spine*, 19(10):1111-1116.

Wasiak, R., Kim, J., and Pransky, G. (2006) Work disability and costs caused by recurrence of low back pain: Longer and more costly than in first episodes. *Spine*, 31(2):219-225.

Wilder, D.G., Aleksiev, A.R., Magnusson, M.L., Pope, M.H., Spratt, K.F., and Goel, V.K. (1996) Muscular response to sudden load: A tool to evaluate fatigue and rehabilitation. *Spine*, 21(22):2628-2639.

APPENDICES

Appendix A: Questionnaires

NOTE: The first 18 questions are the (Body Awareness Questionnaire) BAQ and the subsequent 12 questions are the Kinesthetic and Proprioceptive Awareness Questionnaire (KPAQ).

Listed below are a number of statements related to a variety of normal kinds of feelings and bodily reactions. Read each item and decide how well the statement reflects you personally. It's best to go with your first judgment and not to spend too long mulling over any one question.

Use the response scale provided to choose your answer and mark your answer sheet accordingly

	NEVER true about me	**OCCASIONALLY** true about me	**SOMETIMES** true about me	**FREQUENTLY** true about me	**ALWAYS** true about me
1. I notice differences in the way my body reacts to various foods.	A	B	C	D	E
2. I can always tell when I bump myself whether or not it will become a bruise.	A	B	C	D	E
3. I always know when I've exerted myself to the point where I'll be sore the next day.	A	B	C	D	E
4. I am always aware of changes in my energy level when I eat certain foods.	A	B	C	D	E
5. I know in advance when I'm getting the flu.	A	B	C	D	E
6. I know I'm running a fever without taking my temperature.	A	B	C	D	E
7. I can distinguish between tiredness because of hunger and tiredness because of lack of sleep.	A	B	C	D	E
8. I can accurately predict what time of day lack of sleep will catch up with me.	A	B	C	D	E
9. I am aware of a cycle in my activity level throughout the day.	A	B	C	D	E
10. I *don't* notice seasonal rhythms and cycles in the way my body functions.	A	B	C	D	E
11. As soon as I wake up in the morning I know how much energy I'll have during the day.	A	B	C	D	E
12. I can tell when I go to bed how well I will sleep that night.	A	B	C	D	E
13. I notice distinct body reactions when I am fatigued.	A	B	C	D	E

	NEVER true about me	OCCASIONALLY true about me	SOMETIMES true about me	FREQUENTLY true about me	ALWAYS true about me
14. I notice specific body responses to changes in the weather.	A	B	C	D	E
15. I can predict how much sleep I will need at night in order to wake up refreshed.	A	B	C	D	E
16. When my exercise habits change, I can predict very accurately how that will affect my energy level.	A	B	C	D	E
17. There seems to be a "best" time for me to go to sleep at night.	A	B	C	D	E
18. I notice specific bodily reactions to being over-hungry.	A	B	C	D	E
19. I am aware of my overall body posture.	A	B	C	D	E
20. I am aware of how far I am bending over when I have to bend to do something.	A	B	C	D	E
21. I am aware of strain in my muscles.	A	B	C	D	E
22. I can provide definite information regarding the specific location and severity of pain/discomfort in my body when the doctor asks me what symptoms I am having.	A	B	C	D	E
23. I can exert the correct amount of force/pressure required to do a task even without thinking about it.	A	B	C	D	E
24. I am sensitive to changes in the position of my legs even without looking at them.	A	B	C	D	E
25. I can touch my nose with my index fingers, even with my eyes closed.	A	B	C	D	E
26. I can tell when I should stop doing something (e.g., lifting) before it causes me pain or injury.	A	B	C	D	E
27. I can tell where my hands are located without even looking at them.	A	B	C	D	E
28. I can tell how tired I will be after a task when I first start doing it.	A	B	C	D	E
29. I can feel even the slightest touch (e.g., a small raindrop or an ant crawling) on my skin.	A	B	C	D	E
30. I know my own strength.	A	B	C	D	E

The proprioception survey data, represented as the sum of the scores from the BAQ and the KPAQ surveys, showed a strong correlation with the trunk repositioning precision data from Period 1. This relationship is shown in Figure 44. Since the trunk repositioning data used was from Period 1 before any of the fatiguing task was performed, the data from both the control subjects and the experimental subjects was used. The data showed that the precision for each plane showed a similar negative relationship with the KPAQ+BAQ score. Data for only 8 subjects is shown in the graph since one of the subjects' kinematic data was corrupted. Also, two subjects had the same datapoint (109, 1.4) which show up as only one point on the graph. Table 26 shows the relationship of the proprioception score with the individual plane precision measures. These data suggests that the subjects were able to assess their level of proprioception with some level of accuracy.

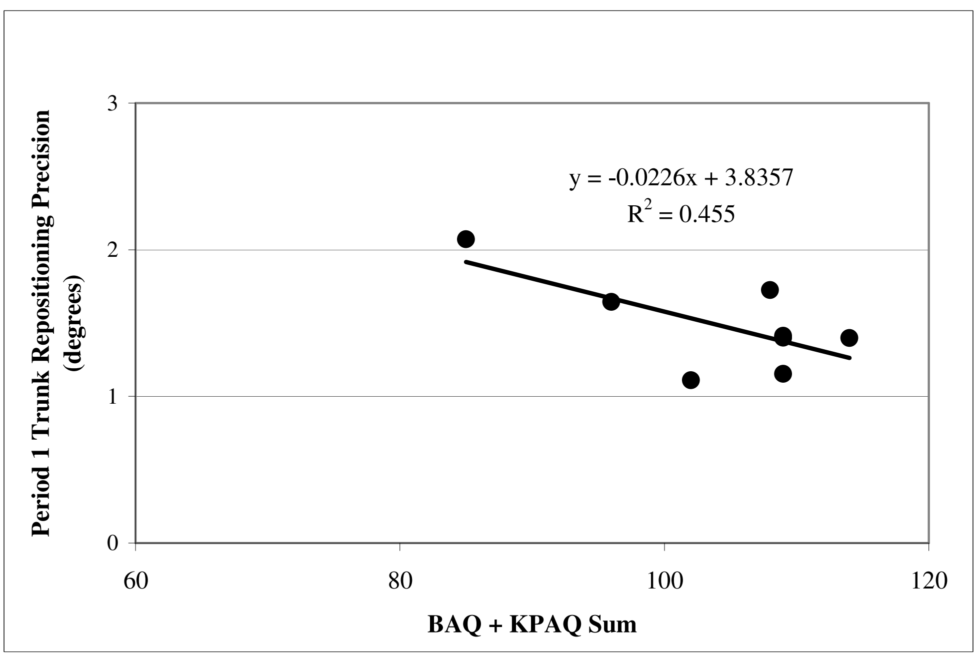

Figure 44: The relationship between the trunk repositioning error variability and the proprioception awareness.

Table 26: Relationships between the trunk repositioning precision and the combined KPAQ+BAQ score. The repositioning precision data is only from Period 1 data for both experimental and control subjects.

Muscle	**R^2 with KPAQ+BAQ**
Flexion	0.178
Lateral Bending	0.290
Axial Rotation	0.031
Average of all 3 planes	**0.455**

Appendix B: Data Collection Session Timeline

Task #	INTRODUCTION	Elapsed Time (min:sec)
1	Introduction & explain experiment	5:00
2	Introduction and sign informed consent form	10:00
3	Stretch	15:00
	PRACTICE TESTS	
4	**A--Box Lift Test**	17:00
5	**B--Trunk Repositioning Test**	19:00
	{Get subject into ARF}	21:00
6	Set up ARF	22:00
7	**C--Force Replication Test**	24:00
8	**D--Muscle Onset Time Test**	26:00
9	**E--Max Extension Test**	27:00
10	Set up ARF	28:00
11	**F--EMG Test, sagittally symmetric**	30:00
12	Set up ARF	31:00
13	**G--EMG Test, asymmetric**	33:00
	SETUP TASKS	
14	Prepare electrodes	38:00
15	Place electrodes on subject in bent-over posture with subject sitting	
16	Check impedances on electrodes	58:00
17	Prep for birds (shave and alcohol)	63:00
18	Place birds on subject (L5/S1, T10, C7, bilat thighs, bilat shank)	68:00
19	Measure location of birds relative to hip and knee joint CORs	70:00
20	Strain relief all cables	75:00
21	Verify cables secured and subject can lift/move	80:00
22	Calibrate sensors for Motion Monitor Software	90:00
23	Turn on video	91:00
	BASELINE, PERIOD 1	
	Baseline Borg Ratings	92:00
24	**A--Trunk Repositioning**--Collect Data	93:00
25	**B—Box Lift Test**--Collect Data	95:00
	{Get subject into ARF}	97:00
26	Set EMG gains and verify signals	119:00
27	Set up ARF for maximum exertions	109:00
	EMG Maximum exertions	121:00
28	Standing upright, resting	
29	30^{o} flexed, static	
30	30^{o} flexed, hanging	

31	30° flexed, ext. max	
32	30° flexed, flex. max	
33	30° flexed, 20° asym, static	
34	30° flexed, 20° asym, ext max	
35	30° flexed, 20° asym, flex max	
36	**C--Force Replication Test**	121:00
37	**D--Muscle Onset Time Test**	123:00
38	Set up ARF for EMG test	124:00
39	**F--EMG Test, sagittally symmetric**	126:00
40	Set up ARF for EMG test	127:00
41	**G--EMG Test, asymmetric**	129:00
	PERIOD 2	
42	**Set up ARF for fatiguing trunk extensions**	130:00
43	**ARF fatiguing task (6 lifts/min, sagittally symmetric, 30 repetitions)**	135:00
	Borg Ratings during fatiguing task	
44	ARF Protocol **1A--EMG tests**, stop=40, start=30	136:00
45	**A--Max Extension 1 rep**--Collect Data	137:00
	{Get subject out of ARF}	139:00
46	**B--Trunk Repo**--Collect Data	142:00
	{Get subject into ARF}	144:00
47	ARF Protocol **1A--EMG tests**, stop=50, start=0	145:00
48	**C--Force Repo**--Collect Data	147:00
49	**D--EMG tests, 0 degrees**--Collect Data	149:00
50	**ARF Protocol 2**--fatigue, stop=50, start=0, 20% MVC	150:00
51	**ARF fatiguing task (6 lifts/min, sag symm, 30 lifts)**	155:00
	Borg Ratings	
52	ARF Protocol **1A--EMG tests**, stop=40, start=30	156:00
53	**E--Max Extension 1 rep**--Collect Data	157:00
	{Get subject out of ARF}	159:00
54	**F--Lift Test**--Collect Data	161:00
	{Get subject into ARF}	163:00
55	**G--Rxn Time**--Collect Data	165:00
56	ARF Protocol **1B--EMG tests**, stop=50, start=0	166:00
57	**H--EMG tests, 20 degrees**--Collect Data	168:00
	PERIOD 3	
58	**ARF Protocol 3**--fatigue, stop=50, start=0, 30% MVC	169:00
59	**ARF fatiguing task (6 lifts/min, sag symm, 30 lifts)**	174:00
	Borg Ratings	
60	ARF Protocol **1A--EMG tests**, stop=40, start=30	175:00
61	**A--Max Extension 1 rep**--Collect Data	176:00
	{Get subject out of ARF}	178:00

62	**B--Trunk Repo**--Collect Data	181:00
	{Get subject into ARF}	183:00
63	ARF Protocol **1A--EMG tests**, stop=50, start=0	184:00
64	**C--Force Repo**--Collect Data	186:00
65	**D--EMG tests, 0 degrees**--Collect Data	188:00
66	**ARF Protocol 3**--fatigue, stop=50, start=0, 30% MVC	189:00
67	**ARF fatiguing task (6 lifts/min, sag symm, 30 lifts)**	194:00
	Borg Ratings	
68	ARF Protocol **1A--EMG tests**, stop=40, start=30	195:00
69	**E--Max Extension 1 rep**--Collect Data	196:00
	{Get subject out of ARF}	198:00
70	**F--Lift Test**--Collect Data	200:00
	{Get subject into ARF}	202:00
71	**G--Rxn Time**--Collect Data	204:00
72	ARF Protocol **1B--EMG tests**, stop=50, start=0	205:00
73	**H--EMG tests, 20 degrees**--Collect Data	207:00
	PERIOD 4	
74	**ARF Protocol 3**--fatigue, stop=50, start=0, 30% MVC	208:00
75	**ARF fatiguing task (6 lifts/min, sag symm, 30 lifts)**	213:00
	Borg Ratings	
76	ARF Protocol **1A--EMG tests**, stop=40, start=30	214:00
77	**A--Max Extension 1 rep**--Collect Data	215:00
	{Get subject out of ARF}	217:00
78	**B--Trunk Repo**--Collect Data	220:00
	{Get subject into ARF}	222:00
79	ARF Protocol **1A--EMG tests**, stop=50, start=0	223:00
80	**C--Force Repo**--Collect Data	225:00
81	**D--EMG tests, 0 degrees**--Collect Data	227:00
82	**ARF Protocol 3**--fatigue, stop=50, start=0, 30% MVC	228:00
83	**ARF fatiguing task (6 lifts/min, sag symm, 30 lifts)**	233:00
	Borg Ratings	
84	ARF Protocol **1A--EMG tests**, stop=40, start=30	234:00
85	**E--Max Extension 1 rep**--Collect Data	235:00
	{Get subject out of ARF}	237:00
86	**F--Lift Test**--Collect Data	239:00
	{Get subject into ARF}	241:00
87	**G--Rxn Time**--Collect Data	243:00
88	ARF Protocol **1B--EMG tests**, stop=50, start=0	244:00
89	**H--EMG tests, 20 degrees**--Collect Data	246:00
	FINISH DATA COLLECTION	
	Remove electrodes	251:00
	Total Time	**251**:00

Appendix C: Subject Instructions

These instructions will be read to the subject for the practice sessions, not for the data collection session.

Muscle Activity Tests and Repetitive Trunk Extensions

The goal of these tests is to push back against the bar with a constant force during dynamic trunk extensions with your eyes open. Once the computer has started, push back against the bar to start the bar moving. Keep pushing against the bar as it goes down and then back up while keeping the bouncing ball force trace in the blue training zone on the screen. If the ball is above the zone, push against the bar with less force until the ball drops into the zone. If the ball is below the zone, push harder against the bar until the ball rises into the zone. Remember to aim to keep the ball in the center of the zone for the entire set of repetitions. You will do this for 10 continuous repetitions for the muscle activity tests. You will do this for 30 sets of one repetition each with 4 seconds of rest between each single-rep set for the repetitive trunk extensions.

Strength Test

The goal of this test is to push against the bar with maximal force. When I say "go", gradually start pushing against the bar, taking about a second or two to ramp up to a maximal force. Hold that force until I say stop. Remember, don't ramp up your speed very quickly, I want to see a maximal force held for over one second.

Muscle Onset Time Test

The goal of this test is to push back against the bar as quickly as you can as soon as you feel the bar push against your back while your eyes are closed. Once the test is started, I will ask you to close your eyes, cross your arms over your chest, and position your body with your back just touching the bar. Then, I will start the bar moving. As soon as you feel the bar moving, quickly push back against the bar. The bar will stop moving automatically once you hit 60% of your maximal force level. After each repetition, we will reset the bar to the starting position and start over. You will do this while bent over 30^{o} forward. You will do 3 repetitions of this for each test. I will randomize the time between when you close your eyes and when I start the test so you can't predict when the bar will start moving.

Force Reproduction Test

The goal of these tests is to replicate an isometric force with your back as accurately as you can 10 times with your eyes closed. First, you will push against the bar so the force level is in the blue training zone. Hold that force there, then close your eyes and release the force. Then, each time the computer beeps, push back against the bar with the exact same force as the first repetition. You will do 10 repetitions with your eyes closed. Remember to ramp up the force over about a second of time, hold that force, then stop pushing when I say stop.

Ratings of Perceived Exertion

The goal of this information is to estimate your perceived exertion levels as accurately as you can each time you are asked. Towards the end of the 30 repetitions, I will ask you to rate your level of perceived exertion for the arms and hands, upper back, lower back, legs, and whole body.

Lifting Test

The goal of this test is to lift and lower a box placing it accurately at its destination. You will be lifting this milk crate from the top of the stack on your right to the tape square on the floor to your left. Then you will lift the box back to the top of the stack to your right again. After each lift and each lower, pause for a second and stand upright, and look straight ahead at the far wall. Then start your next lift. Lift with your own lifting style and speed, but don't move your feet. Place the box as accurately as you can in the target location without making any fine adjustments after you put it down.

Trunk Repositioning Test

The goal of these tests is to replicate a posture as accurately as you can 10 times with your eyes closed. First, bend over forward with your hands clasped together until your fingertips touch the top of the box. Then, in that exact position, close your eyes, cross your arms across your chest, then stand straight up. Then, each time the computer beeps, bend forward to the exact position as the first repetition. You will do 10 repetitions with your eyes closed. Remember to keep your neck as straight as possible, do not bend your neck relative to your trunk.

Immediately after you finish the forward trunk repositioning repetitions, you will start the side bending trunk repositioning tests. With your eyes open, bend to the side until your fingertips touch the top of the box. Then close your eyes and cross your arms and stand straight up. Then, each time the computer beeps, bend to the side to the exact position as the first repetition. You will do 10 repetitions with your eyes closed.

Immediately after you finish the side bending trunk repositioning repetitions, you will start the twisting trunk repositioning tests. With your eyes open and your arms up to the sides, twist to the right until your right arm touches the wood piece behind you. Then close your eyes and cross your arms and return to facing forward. Then, each time the computer beeps, bend to the side to the exact position as the first repetition. You will do 10 repetitions with your eyes closed. Remember to only twist at the spine, keep your hips facing forward the entire test.

Appendix D: Verification of ANOVA Assumptions

Muscle Activity Test

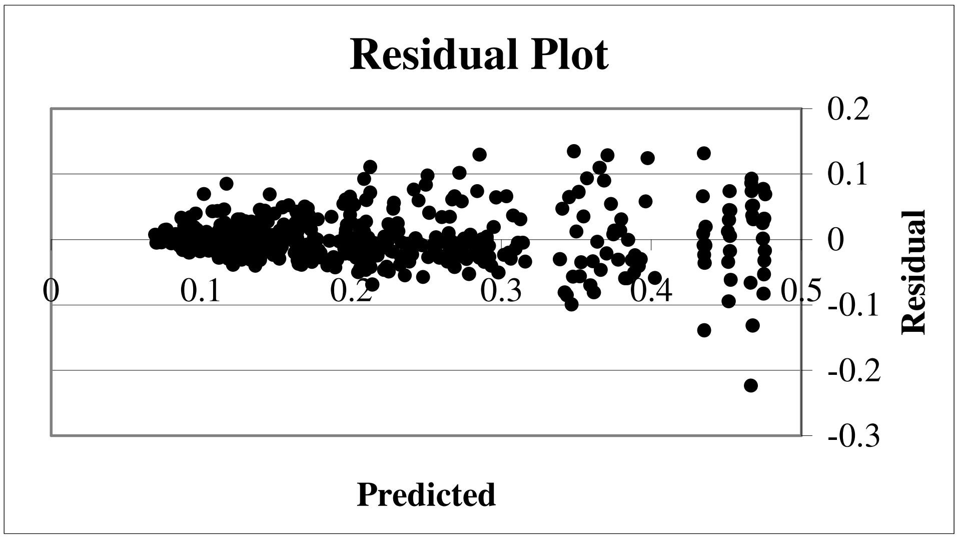

Figure 45: Residual plot of the untransformed NEMG of the Left External Oblique (medial).

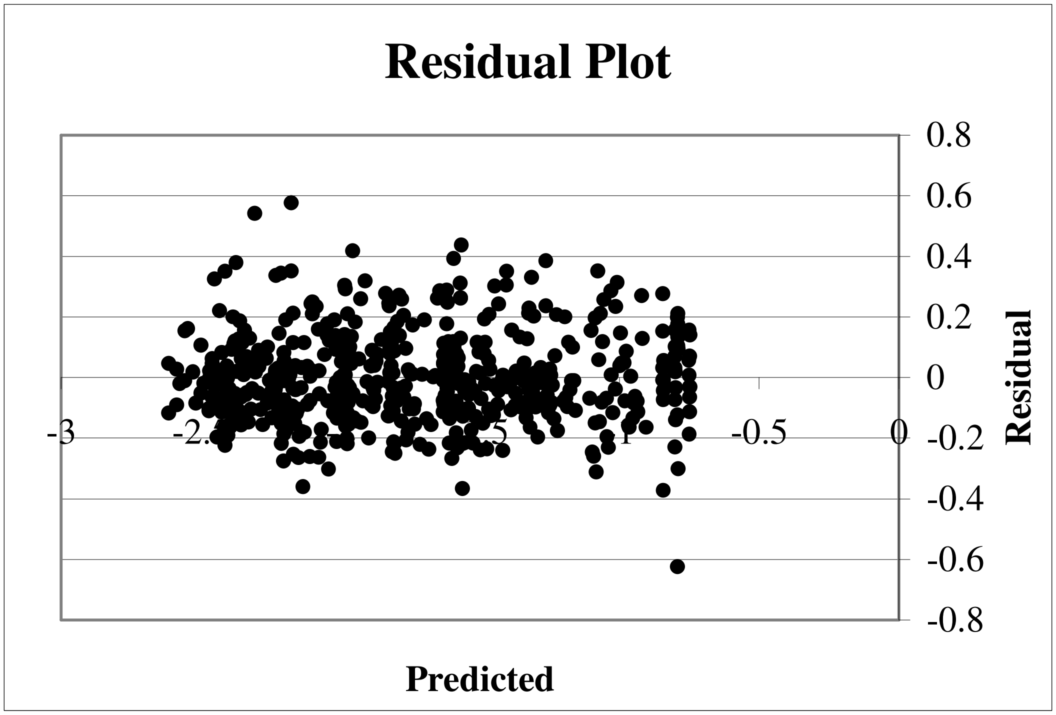

Figure 46: Residual plot of the natural log of the Left External Oblique (medial).

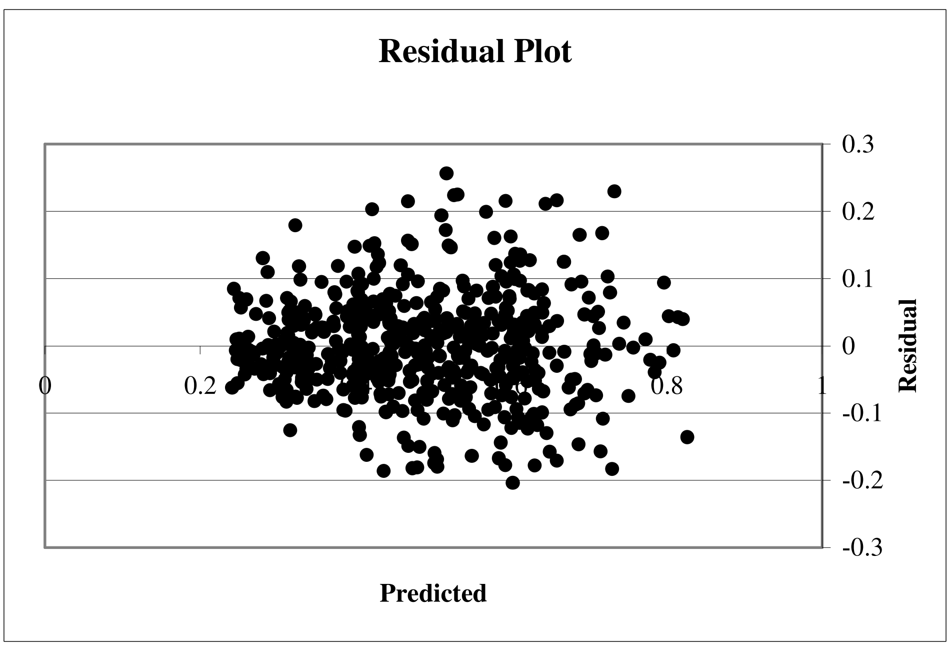

Figure 47: Residual plot of the Right Longissimus.

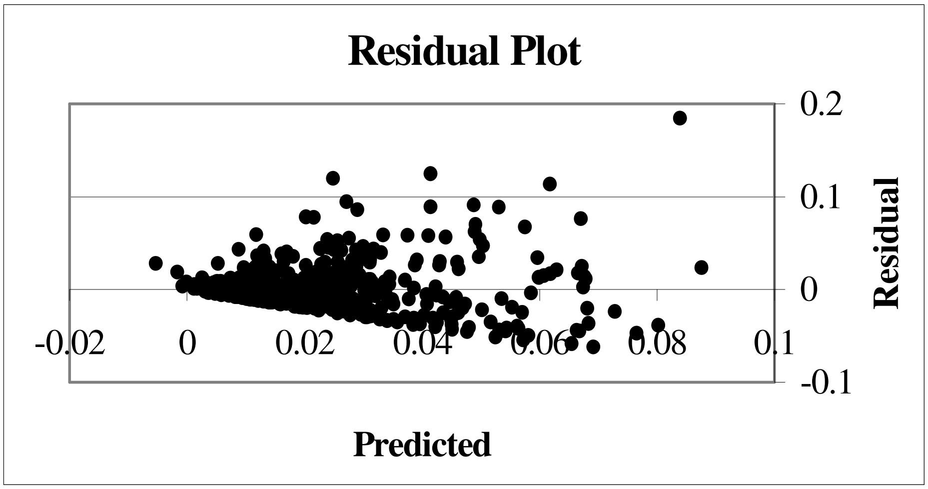

Figure 48: Residual plot of the untransformed Levene's measures of the Left External Oblique (medial).

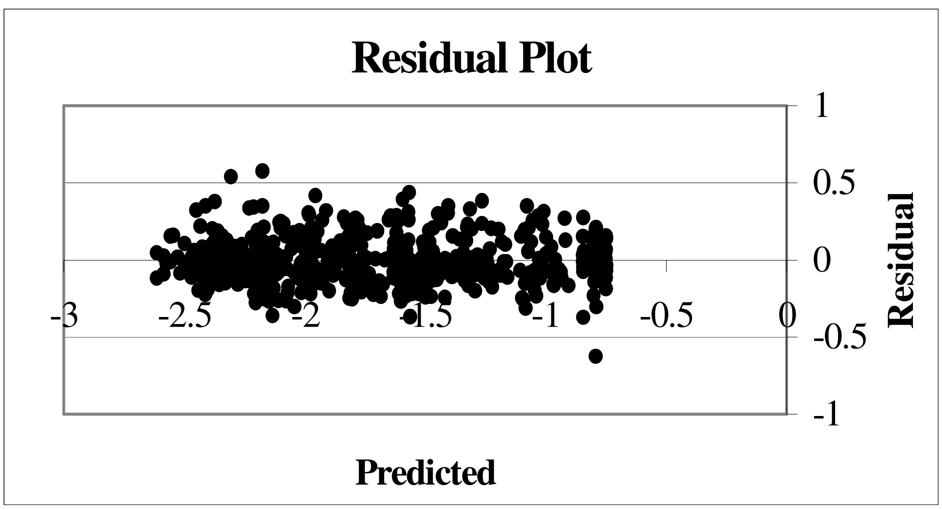

Figure 49: Residual plot of the natural log-transformed Levene's measures of the Left External Oblique (medial).

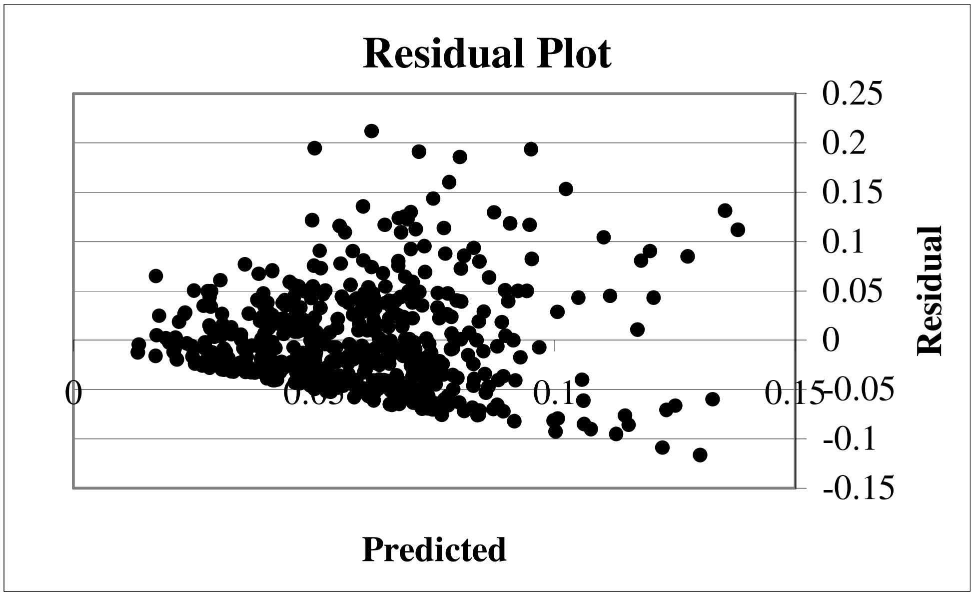

Figure 50: Residual plot of the untransformed Levene's measures of the Right Longissimus.

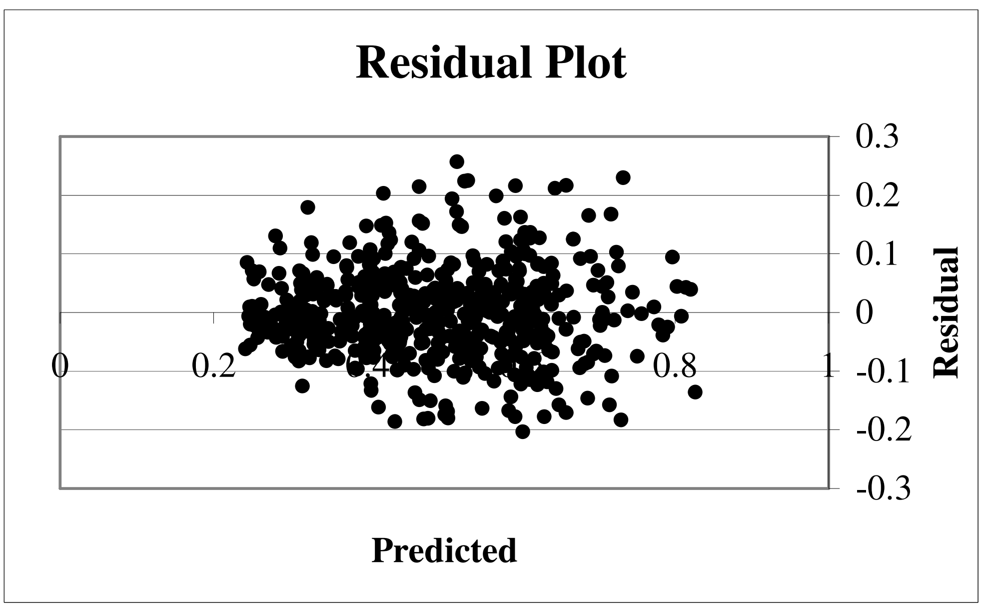

Figure 51: Residual plot of the natural log-transformed Levene's measures of the Right Longissimus.

Force Replication Test

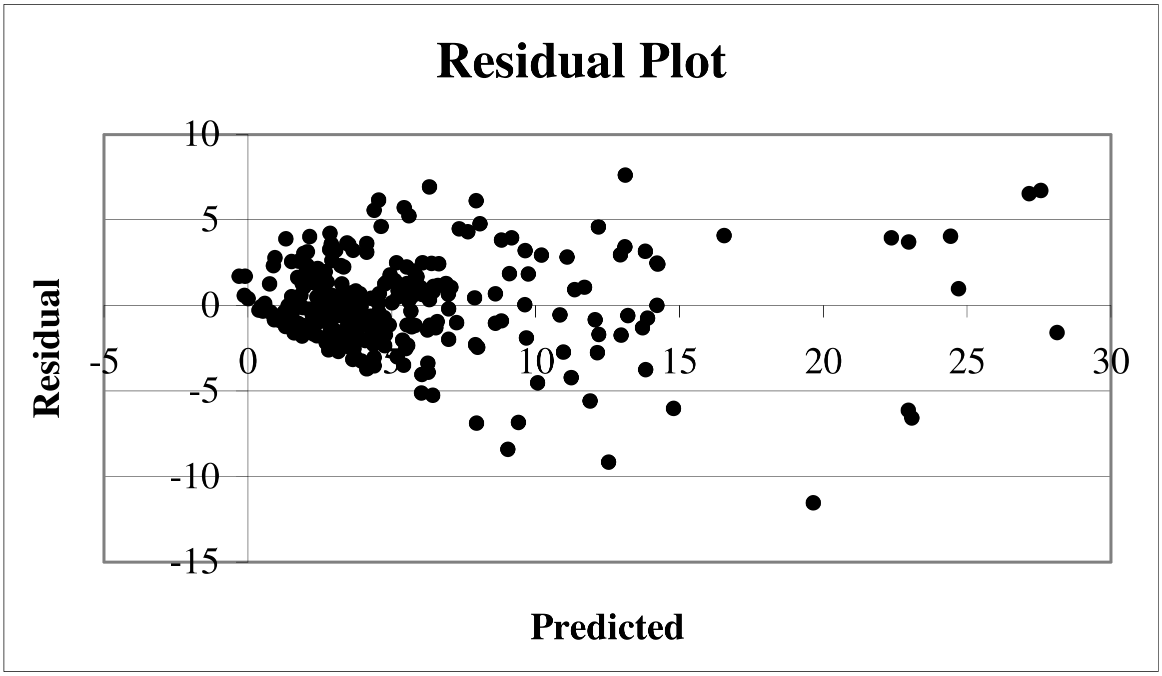

Figure 52: Untransformed Levene's test for force replication error.

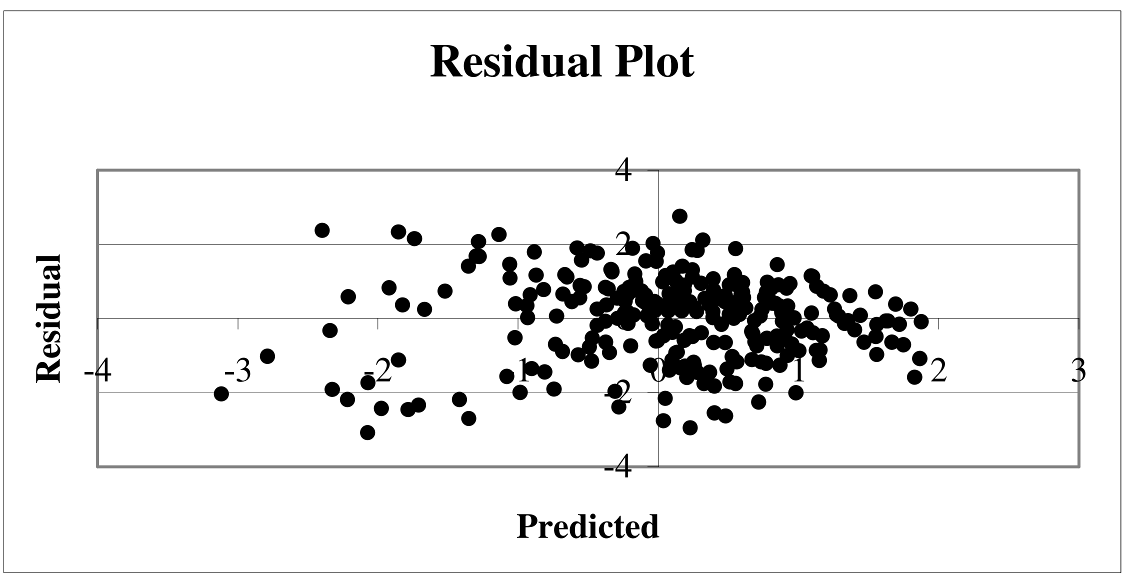

Figure 53: Log transformed Levene's test for force replication error.

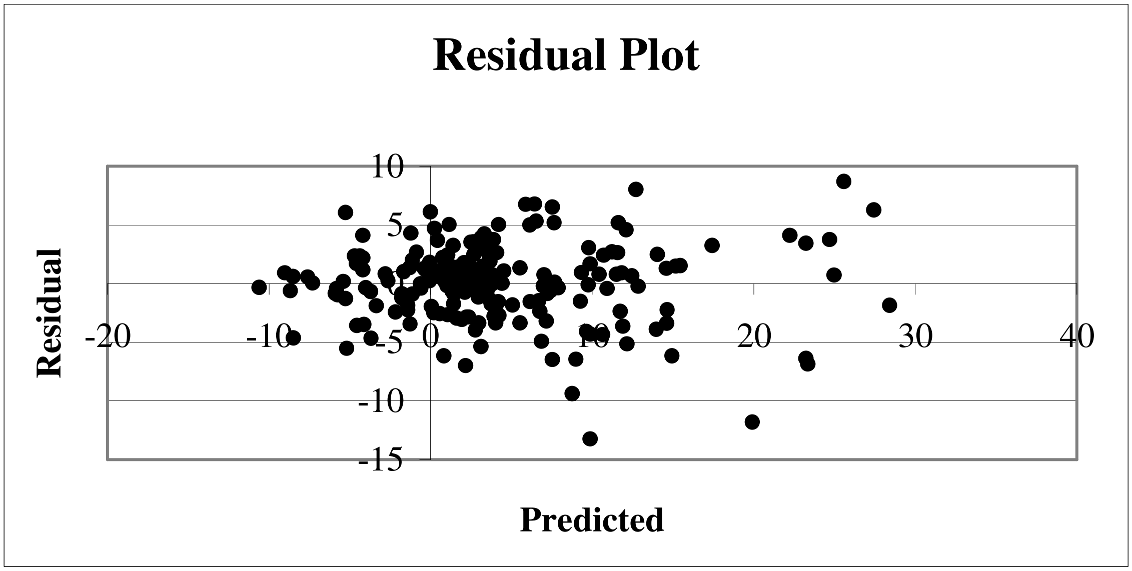

Figure 54: Force replication error residual plot.

Muscle Onset Test

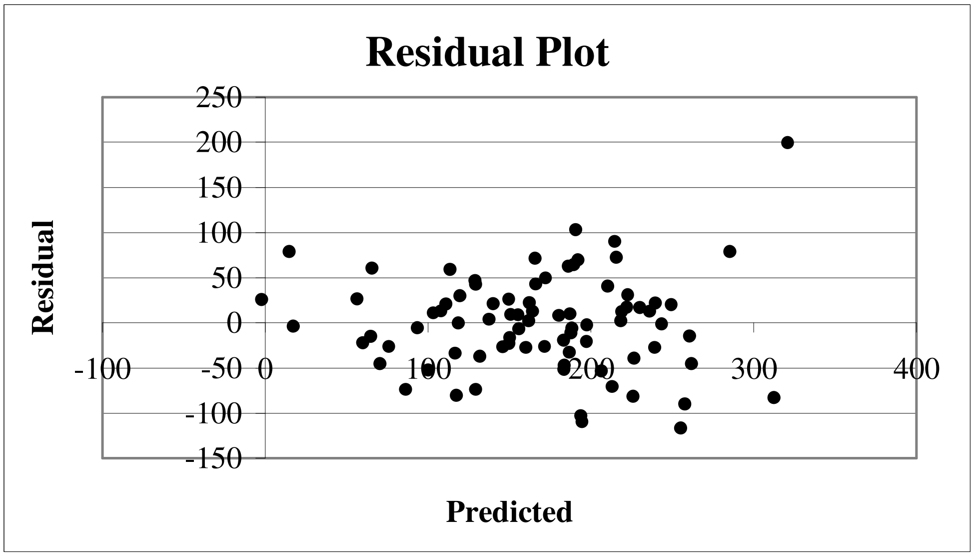

Figure 55: Residual plot for the untransformed muscle onset time for the Right Longissimus muscle.

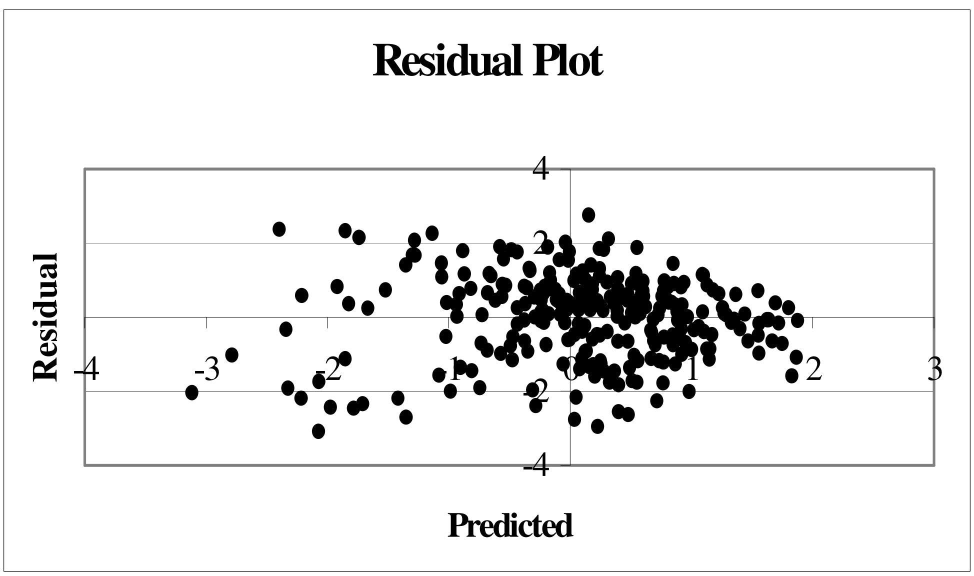

Figure 56: Log transformed Levene's test for force replication error.

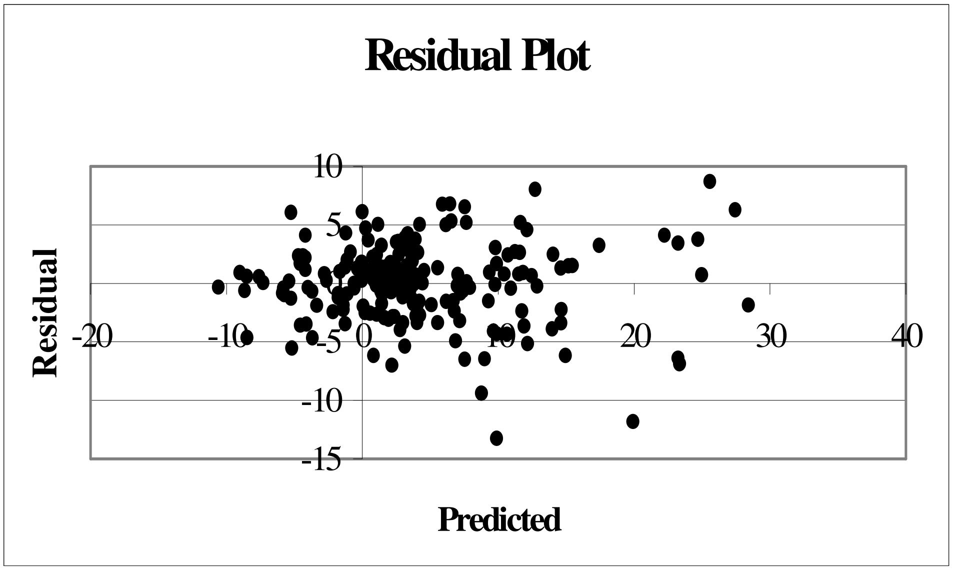

Figure 57: Force replication error residual plot.

Trunk Kinematics Test

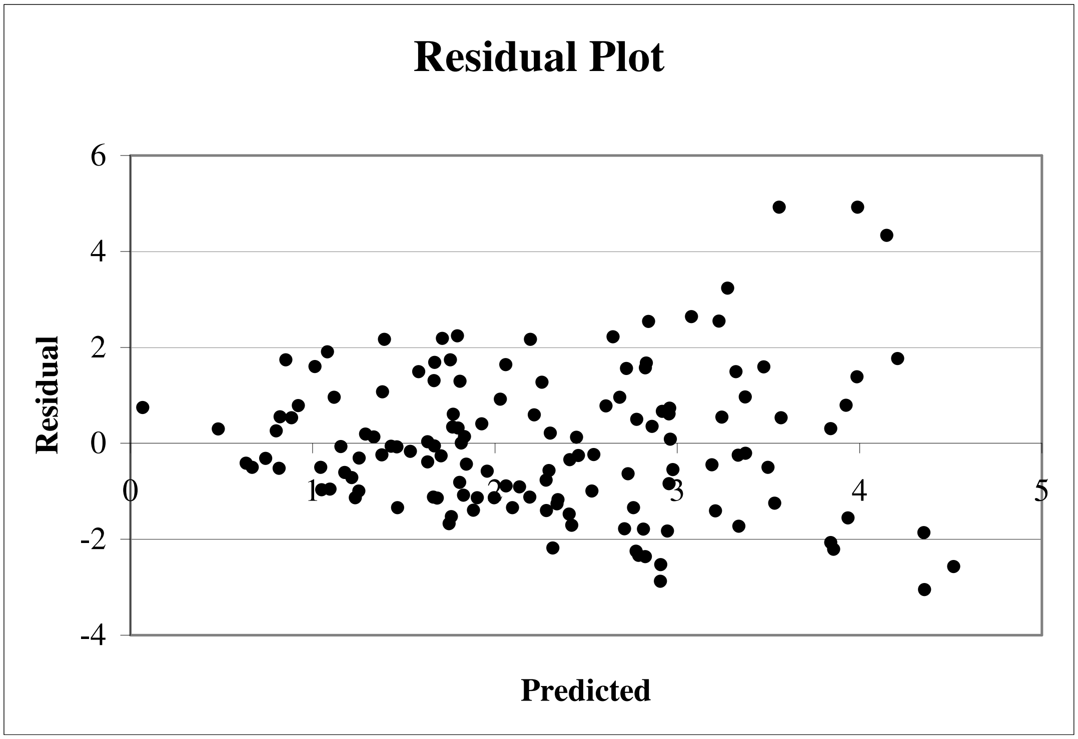

Figure 58: Untransformed Levene's test for trunk flexion during the instant of box liftoff during the trunk kinematics test.

Appendix E: Explanation of Force Replication Test Outlier

Subject number eight in the experiment exhibited an extremely high average absolute error for the trunk force replication test in Period 1. The review of the raw data file showed no anomalies in the data collection or data processing of this file. Figure 59 shows the large effect this one subject had on this dataset. His average absolute error value for Period 1 was more than 6 standard deviations from the AE of all the Period 1 data for the experimental group. Due to this large discrepancy, and because the removal of this subject did not impact the overall statistical analysis of the data, this data was removed from this test for all periods. Lastly, Figure 60 shows the data from all the individual experimental subjects and the datapoint from Subject 8 is obviously an outlier.

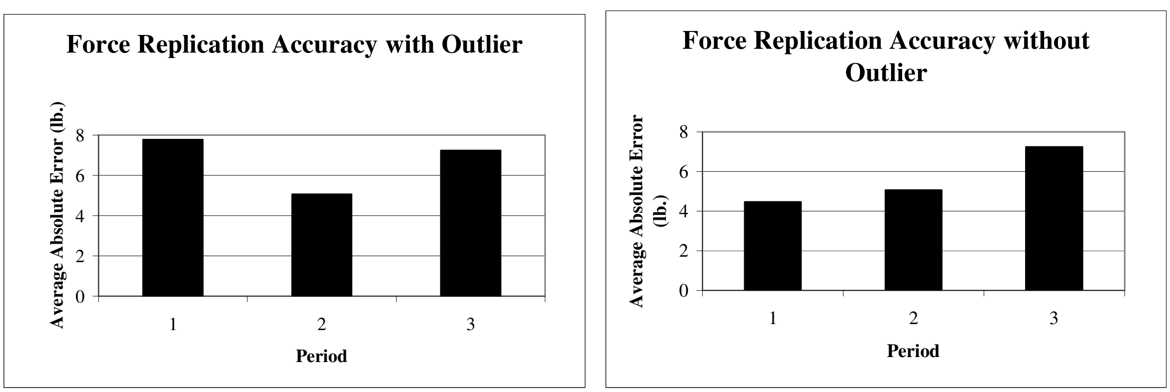

Figure 59. The AE with (left) and without (right) the subject 8 datapoint from Period 1.

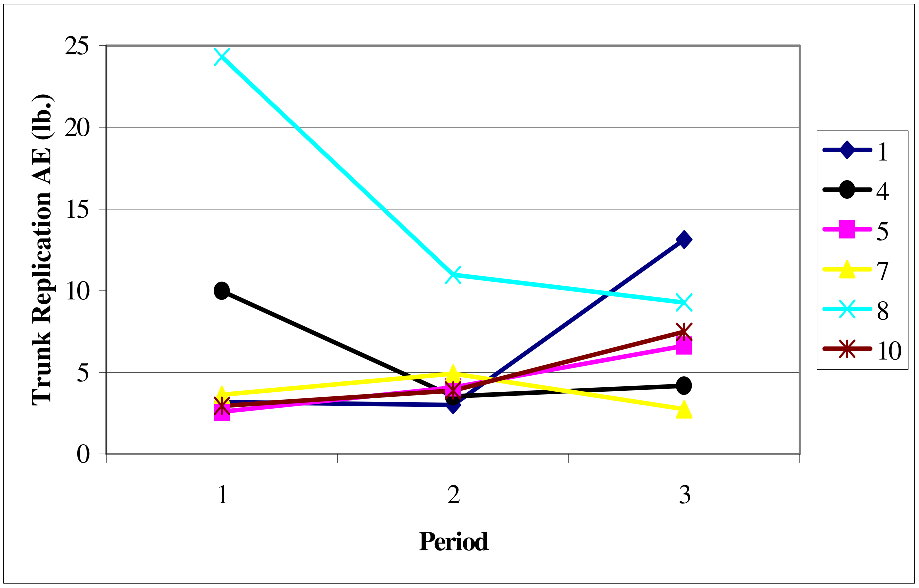

Figure 60: The AEs for each individual in the experimental group including the outlier data from subject 8.

Appendix F: Individual Subject Trends

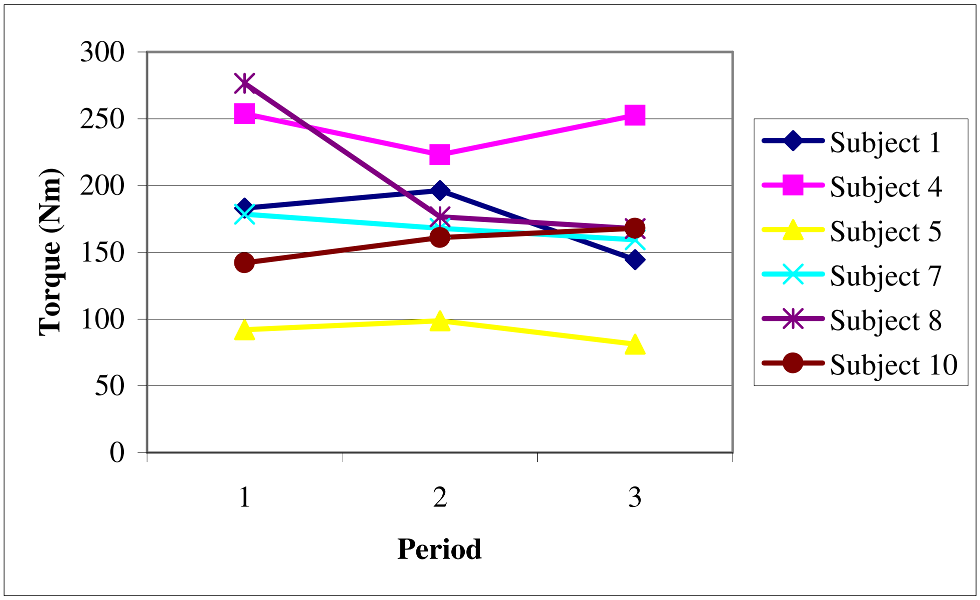

Figure 61: The individual trends for maximum trunk torque.

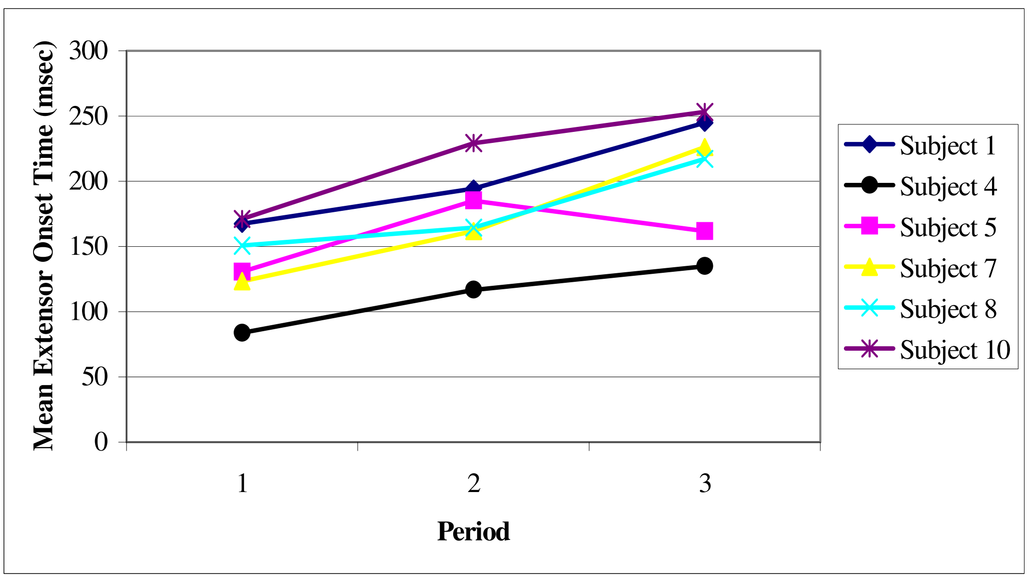

Figure 62: The individual trends for trunk extensor muscle onset time. Each datapoint is averaged across the four extensor muscles.

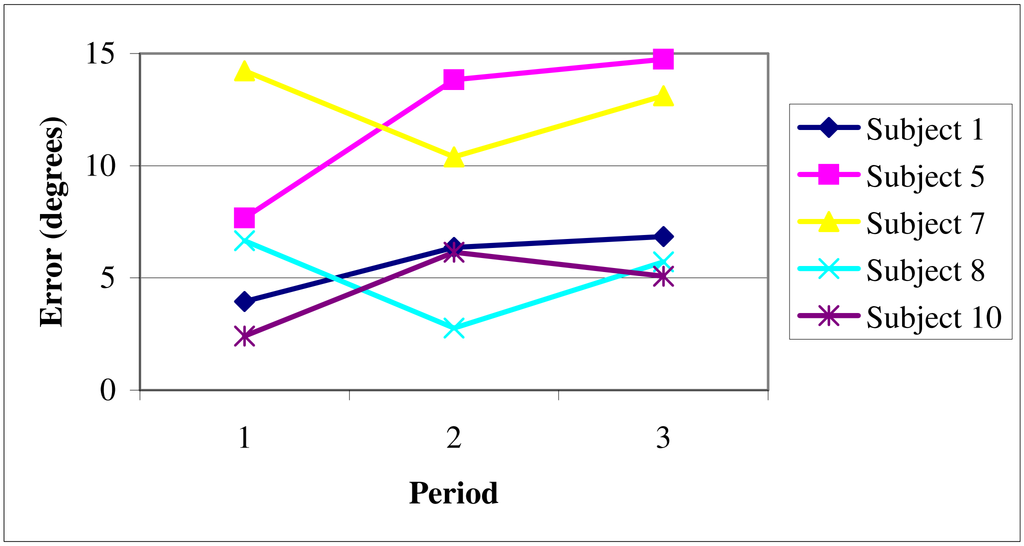

Figure 63: The individual trends for trunk repositioning accuracy during the flexion trials.

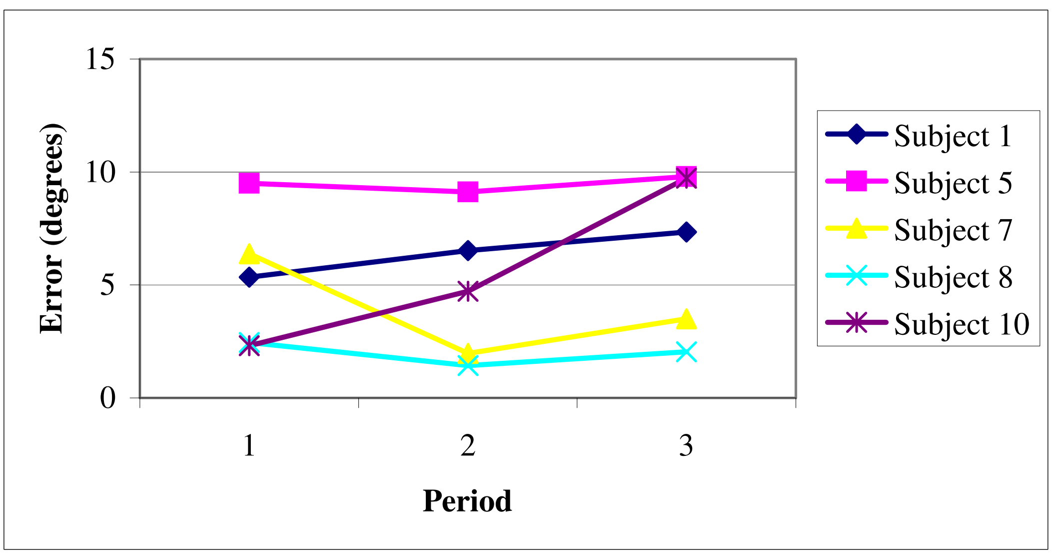

Figure 64: The individual trends for trunk repositioning accuracy during the lateral bending trials.

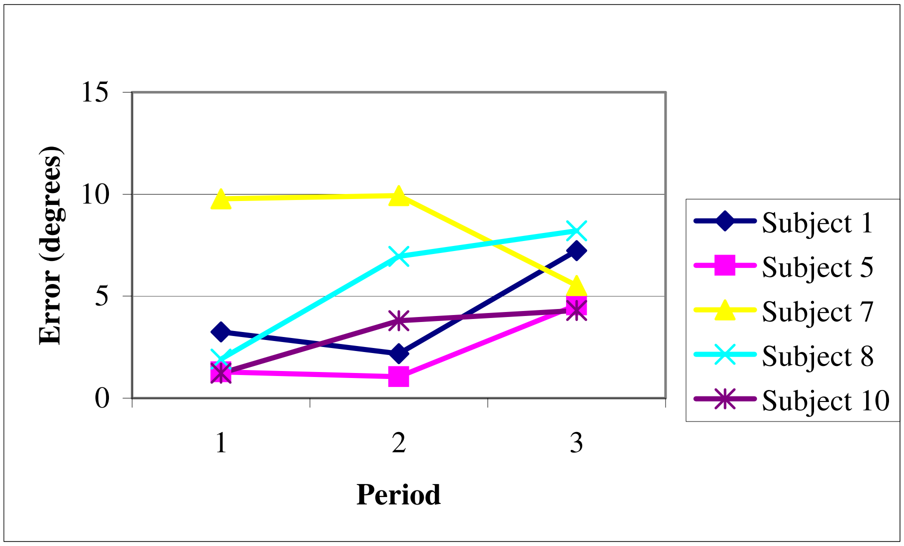

Figure 65: The individual trends for trunk repositioning accuracy during the axial rotation trials.

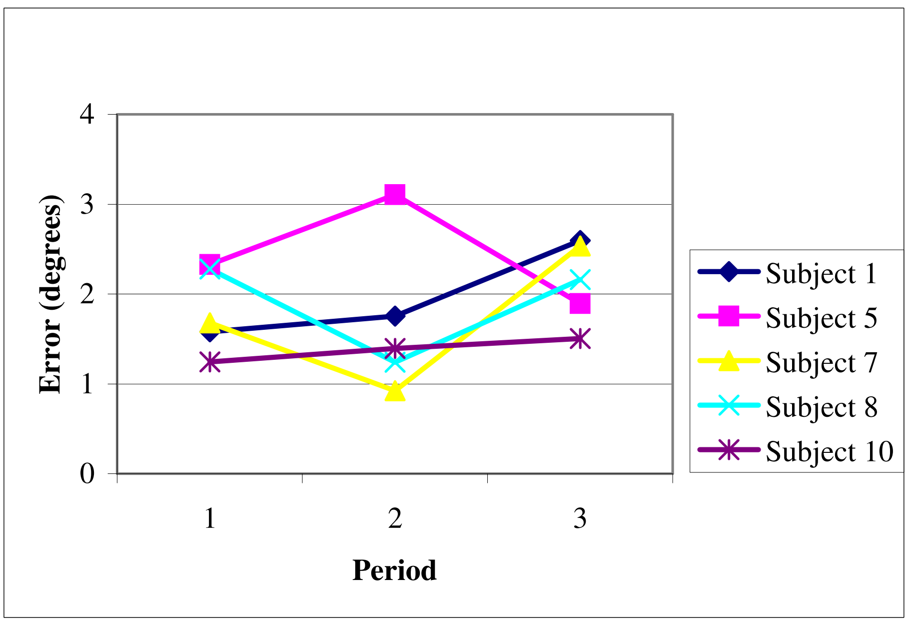

Figure 66: The individual trends for trunk repositioning precision during the flexion trials.

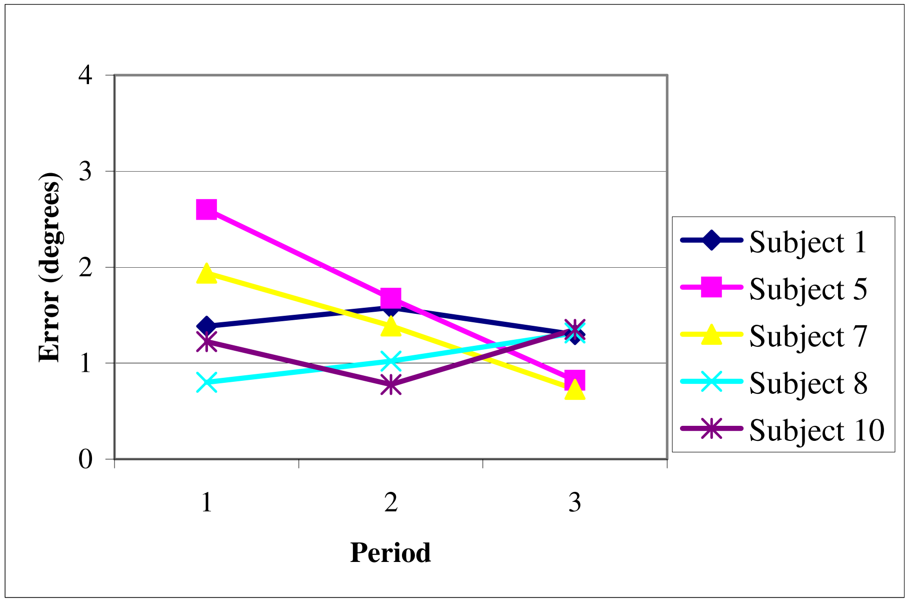

Figure 67: The individual trends for trunk repositioning precision during the lateral bending trials.

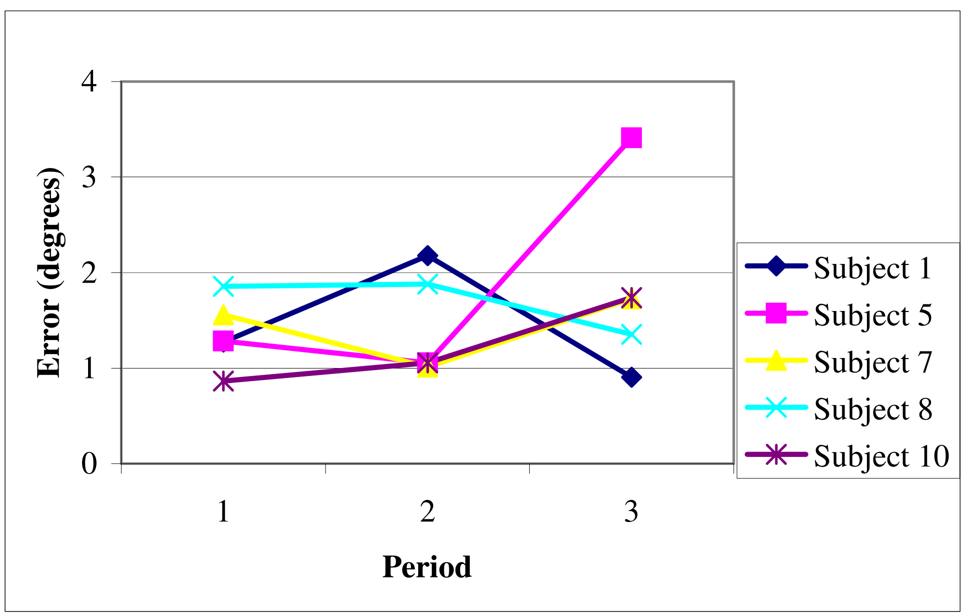

Figure 68: The individual trends for trunk repositioning precision during the axial rotation trials.

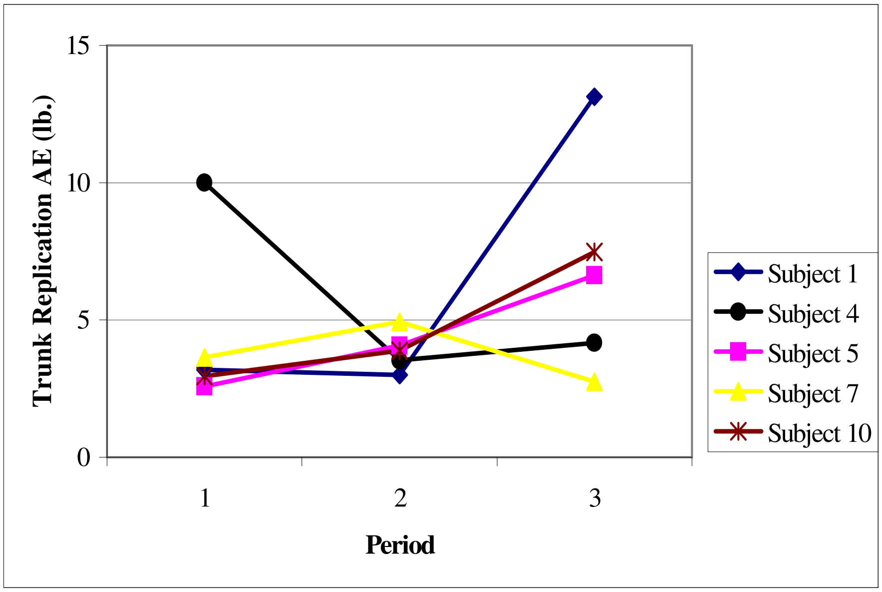

Figure 69: The individual trends for trunk force replication.

Appendix G: Full Dataset Correlation Matrices

Table 27: Correlation matrix for full dataset.

	Torque	avg0flex	avg0ext	avg20flex	avg20ext	var0flx	var0ext	var20flx	var20ext	frep_abs	frep_var	repo_avg	repo_var
torque	1.000	0.028	-0.096	0.141	-0.224	-0.008	-0.133	-0.168	**-0.721**	0.195	0.479	0.389	0.261
avg0flex	0.028	1.000	-0.039	**0.886**	-0.183	**0.771**	0.165	0.557	0.292	0.010	0.194	0.185	0.059
avg0ext	-0.096	-0.039	1.000	0.122	**0.772**	-0.430	0.412	0.110	-0.210	0.350	0.147	0.453	0.235
avg20flex	0.141	**0.886**	0.122	1.000	0.045	0.607	0.356	**0.744**	0.162	0.219	0.297	0.200	0.011
avg20ext	-0.224	-0.183	**0.772**	0.045	1.000	-0.315	0.557	0.182	0.162	0.058	-0.305	0.130	0.063
var0flx	-0.008	**0.771**	-0.430	0.607	-0.315	1.000	0.095	0.559	0.504	-0.285	-0.135	-0.019	0.112
var0ext	-0.133	0.165	0.412	0.356	0.557	0.095	1.000	0.586	0.203	0.488	0.234	0.276	0.492
var20flx	-0.168	0.557	0.110	**0.744**	0.182	0.559	0.586	1.000	0.323	0.206	0.139	0.172	0.135
var20ext	**-0.721**	0.292	-0.210	0.162	0.162	0.504	0.203	0.323	1.000	-0.490	-0.617	-0.461	-0.127
frep_abs	0.195	0.010	0.350	0.219	0.058	-0.285	0.488	0.206	-0.490	1.000	**0.800**	0.502	0.333
frep_lv	0.479	0.194	0.147	0.297	-0.305	-0.135	0.234	0.139	-0.617	**0.800**	1.000	**0.704**	0.545
repo_avg	0.389	0.185	0.453	0.200	0.130	-0.019	0.276	0.172	-0.461	0.502	**0.704**	1.000	**0.691**
repo_lv	0.261	0.059	0.235	0.011	0.063	0.112	0.492	0.135	-0.127	0.333	0.545	**0.691**	1.000
rxn_avg	-0.095	**-0.667**	0.396	-0.476	0.498	**-0.711**	0.091	-0.211	-0.224	0.098	-0.022	0.168	0.025
lumflexavg	0.453	-0.359	-0.394	-0.507	-0.611	-0.188	-0.573	-0.600	-0.479	-0.181	0.171	0.181	0.216
lumlatavg	0.600	0.487	-0.072	0.444	-0.314	0.319	-0.334	-0.055	-0.301	-0.146	0.143	0.200	-0.031
lumlatvelavg	-0.389	-0.471	-0.332	-0.515	-0.217	-0.282	-0.148	-0.184	0.152	-0.124	-0.126	-0.274	-0.108
lumrotvelavg	-0.361	-0.036	-0.096	-0.001	0.083	0.157	0.628	0.467	0.379	0.284	0.212	0.150	0.453
lumflexvelavg	0.024	-0.605	-0.191	-0.628	-0.234	-0.507	-0.222	-0.461	-0.266	0.013	0.166	0.012	0.156
kneeflexavg	0.331	0.537	-0.305	0.644	-0.199	**0.669**	0.595	**0.693**	0.106	0.320	0.382	0.166	0.321
lumflexvar	0.243	0.623	-0.261	**0.716**	-0.236	0.592	0.484	**0.739**	0.175	0.160	0.455	0.307	0.328
lumlatvar	-0.215	0.030	0.584	0.086	0.623	-0.222	0.220	-0.061	0.223	-0.014	-0.148	-0.101	-0.039
lumlatvelvar	**-0.693**	-0.284	-0.184	-0.450	-0.123	-0.148	-0.486	-0.231	0.337	-0.352	-0.554	-0.364	-0.495
lumrotvelvar	**-0.718**	0.342	0.212	0.188	0.087	0.232	0.198	0.409	0.607	-0.180	-0.079	0.058	0.159

Table 28: Correlation matrix for full dataset (continued).

	rxn_avg	lumflexavg	lumlatavg	lumlatvelavg	lumrotvelavg	lumflexvelavg	kneeflexavg	lumflexvar	lumlatvar	lumlatvelvar	lumrotvelvar
torque	-0.095	0.453	0.600	-0.389	-0.361	0.024	0.331	0.243	-0.215	**-0.693**	**-0.718**
avg0flex	**-0.667**	-0.359	0.487	-0.471	-0.036	-0.605	0.537	0.623	0.030	-0.284	0.342
avg0ext	0.396	-0.394	-0.072	-0.332	-0.096	-0.191	-0.305	-0.261	0.584	-0.184	0.212
avg20flex	-0.476	-0.507	0.444	-0.515	-0.001	-0.628	0.644	**0.716**	0.086	-0.450	0.188
avg20ext	0.498	-0.611	-0.314	-0.217	0.083	-0.234	-0.199	-0.236	0.623	-0.123	0.087
var0flx	**-0.711**	-0.188	0.319	-0.282	0.157	-0.507	**0.669**	0.592	-0.222	-0.148	0.232
var0ext	0.091	-0.573	-0.334	-0.148	0.628	-0.222	0.595	0.484	0.220	-0.486	0.198
var20flx	-0.211	-0.600	-0.055	-0.184	0.467	-0.461	**0.693**	**0.739**	-0.061	-0.231	0.409
var20ext	-0.224	-0.479	-0.301	0.152	0.379	-0.266	0.106	0.175	0.223	0.337	0.607
frep_abs	0.098	-0.181	-0.146	-0.124	0.284	0.013	0.320	0.160	-0.014	-0.352	-0.180
frep_lv	-0.022	0.171	0.143	-0.126	0.212	0.166	0.382	0.455	-0.148	-0.554	-0.079
repo_avg	0.168	0.181	0.200	-0.274	0.150	0.012	0.166	0.307	-0.101	-0.364	0.058
repo_lv	0.025	0.216	-0.031	-0.108	0.453	0.156	0.321	0.328	-0.039	-0.495	0.159
rxn_avg	1.000	0.273	**-0.651**	0.627	0.315	**0.726**	-0.474	-0.311	-0.045	0.299	0.036
lumflexavg	0.273	1.000	0.083	0.425	-0.094	**0.729**	-0.259	-0.207	**-0.667**	0.173	-0.326
lumlatavg	**-0.651**	0.083	1.000	**-0.781**	**-0.762**	-0.552	0.140	0.169	0.035	-0.460	-0.407
lumlatvelavg	0.627	0.425	**-0.781**	1.000	0.564	**0.859**	-0.273	-0.153	-0.460	0.628	0.276
lumrotvelavg	0.315	-0.094	**-0.762**	0.564	1.000	0.376	0.409	0.433	-0.203	0.058	0.482
lumflexvelavg	**0.726**	**0.729**	-0.552	**0.859**	0.376	1.000	-0.348	-0.218	-0.481	0.357	0.008
kneeflexavg	-0.474	-0.259	0.140	-0.273	0.409	-0.348	1.000	**0.820**	-0.204	-0.622	-0.112
lumflexvar	-0.311	-0.207	0.169	-0.153	0.433	-0.218	**0.820**	1.000	-0.159	-0.541	0.200
lumlatvar	-0.045	**-0.667**	0.035	-0.460	-0.203	-0.481	-0.204	-0.159	1.000	-0.280	0.243
lumlatvelvar	0.299	0.173	-0.460	0.628	0.058	0.357	-0.622	-0.541	-0.280	1.000	0.327
lumrotvelvar	0.036	-0.326	-0.407	0.276	0.482	0.008	-0.112	0.200	0.243	0.327	1.000