The Kinematics of the Lumbar Spine during a Ball Bridge Task to Volitional Fatigue in Highly Trained Athletes

by

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A Thesis submitted to the Faculty of Graduate Studies of

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Definitions

Robust-involving or requiring great physical strength and stamina

Neutral Spine-the anatomical position of the spine which includes a lordotic curve at the cervical spine, a kyphotic curve at the thoracic spine and lordotic curve at the lumbar spine-and maintaining this position during physical tasks.

Bridging-the action of maintaining a stiff trunk in a prone or supine position with the trunk elevated from the ground with hands or elbows and feet on the ground. A ball can be used to create a less stable environment

Trunk- the main part of the body, excluding the head, neck, and limbs

Ball Plank Task-A bridge position with the elbows on a stability ball, with the upper arm perpendicular to the floor, and the feet on the ground. The subject then moves the arm forward and then back to start position for repetitions.

Stability-the condition of being stable or the ability to return to its original state after disturbance

Vertebral inclination- the change in angular position in the L1 and L4 vertebrae as represented by the accelerometers affixed onto the spinous processes

Vertebral orientation- the difference in L1 and L4 inclination

Kinematics-the study of motion - a branch of physics that deals with the motion of a body or system without reference to force and mass

Failure- a breakdown or decline in the performance of something, or an occasion when something stops working or stops working adequately

Introduction

Back injuries are very common and back pain affects most people at some point in their lives (Herrmann, Madigan et al. 2006). Postural control and trunk stabilization exercises are thought to be methods of controlling the spine during tasks, are used for injury prevention and are used as rehabilitation methods (Moorhouse and Granata 2005; Lee, Rogers et al. 2006; Reeves, Narendra et al. 2007). It is unclear however, how the muscles surrounding the spine fail and what exercise methods are the most effective in injury prevention and rehabilitation (Barr, Griggs et al. 2005; Koumantakis, Watson et al. 2005). It is also unclear as to how the spine position changes as fatigue occurs from person to person. A better understanding of the motor control and kinematics of spine failure could lead to better injury prevention and rehabilitation methods.

Trunk stabilization exercises have been a main focus in training for high performance athletes as well as a rehabilitation staple for those with back injuries (Standaert, Herring et al. 2004; Barr, Griggs et al. 2005). Despite the belief that strong spine stabilizers are important in optimal function, there is very little information regarding the kinematics of fatigue or failure during trunk stabilization. Some studies have highlighted the effectiveness of trunk muscle activation during certain trunk exercises, such as bridging or plank movements, yet others determine that the effectiveness of general exercise programs are just as great (Koumantakis, Watson et al. 2005; Schellenberg, Lang et al. 2007). Recent research suggests that unstable surfaces and perturbations during training can enhance the ability to activate trunk stabilizers yet it is uncertain as to what occurs directly prior to failure to the position of the spine (Anderson and Behm 2005; Barr, Griggs et al. 2005; Behm, Leonard et al. 2005). Furthermore, these studies have mostly used mostly electromyography (EMG) to determine activation of trunk stabilizers, yet accelerometry could provide added information regarding movement and activation strategy during stabilization exercises. Since back injuries affect so many, it would seem relevant to understand the intricacies of what affects failure from one person to the next.

Athletes are expected to be the example of strength and optimal performance. Many studies have explored trunk stabilization activation in normal subjects or low back pain patients but not highly trained athletes. Understanding the kinematics of fatigue in highly trained athletes can provide some insight as to how training may or may not affect the outcome of the ability to control trunk stabilizers. Highly trained athletes often have more body awareness and control than the rest of the population but also are often subjected to greater ranges of motion and trauma to the spine. Perhaps certain outcomes of fatigue are inevitable whether the subject is a highly trained athlete, low back pain patient or normal subject. This study aims to better understand the kinematics of back failure, using accelerometry during a fatiguing task on an unstable surface, in highly trained athletes from a motor control and lumbar inclination perspective.

Review of Literature

Anatomy of Trunk Stabilization

Knowing the anatomical structures of the trunk is key to understand the complexity of the elements necessary to stabilize the spine. Since the trunk has multiple joints and multiple directional ranges of motion, there are many large and small muscle groups to organize the complex functional requirements of the region. Akuthota and Nadler refer to the trunk or core as the powerhouse or foundation of all limb movements and note the osseous, ligamentous, fascial and muscular structures that the core is made up of. The vertebrae and ligaments as well as the intervertebral disks provide passive stiffness. They list the following as dynamic, phasic, torque producing global muscles: rectus abdominus, external obliques, internal obliques, and thoracis portion of iliocostalis. They note the multifidi, transverse abdominus, quadratus lumborum, diaphragm, psoas major, internal obliques and iliocostalis (lumbar portion) as postural, segmental stabilizing local muscles (Akuthota and Nadler 2004). Each of the muscles has different roles in moving or stabilizing a functional spine.

The thoracolumbar fascia consists of layers that act as a belt in the trunk region providing abdominal attachment sites, proprioceptive feedback, as well as the link between the lower limb and the upper limb(Vleeming, Pool-Goudzwaard et al. 1995). Deep to the thoracolumbar fascia are the paraspinal muscles which include the erector spinae group that act to extend the spine. The local smaller muscles such as multifidi and rotators are theorized to be segment stabilizers and are often atrophied in low back pain patients (Hides, Richardson et al. 1996). The quadratus lumborum, a large thin quadrangular shaped muscle attached to the lumbar spine, is considered to be a major stabilizer and responsible for some lumbar side flexion (Akuthota and Nadler 2004). The fascia, local stabilizers and quadratus lumborum are key to stabilizing the spine yet often over looked in exercise programs.

The abdominals, including the internal obliques, external obliques, and transverse abdominus connected to the thoracolumbar fascia create a hoop around the midsection that serves to functionally stabilize with intra-abdominal pressure. The external obliques are the largest and most superficial abdominals and have multiple roles since it acts as to rotate the trunk, keep the pelvis from excessive anterior rotation and contracts eccentrically in spine extension (Akuthota and Nadler 2004). The rectus abdominus along the anterior abdominal wall acts to flex the spine and can be over emphasized in training since its importance in stabilizing the spine is far less than the transverse abdominus and external obliques (Hodges and Richardson 1996). The diaphragm and the pelvic floor serve to complete the core area with the diaphragm contributing to intra-abdominal pressure upon contraction and the pelvic floor co-contracting during transverse abdominus activation (Akuthota and Nadler 2004). Although the entire abdominal wall is important in the functioning of the trunk-it is only part of the anatomical equation that allows dynamic and functional movement.

The hip musculature is very important to stabilize the pelvis and trunk as well as transfer energy from the lower extremities to the pelvis and spine (Akuthota and Nadler 2004). The Gluteus maximus is a large hip extensor and the gluteus medius stabilizes the pelvis, abducts the hip and medially rotates the hip. Delayed or weak firing of the gluteus maximus or gluteus medius has been related to a greater occurrence of low back pain (Nadler, Malanga et al. 2001). The psoas major acts primarily as a hip flexor however, its proximal attachment is at the anterior lumbar spine potentially making it influential in spinal stability in hip flexed positions (Akuthota and Nadler 2004). Tight psoas muscles could influence the load on the spine and the need for greater trunk stabilization (Juker, McGill et al. 1998; Nadler, Malanga et al. 2001). The entire anatomical description of the trunk is very extensive and complex, yet an awareness of all of the roles involved is important to the understanding of the inter-relationship between the muscle groups, connective tissues and joint support for daily optimal functioning.

Trunk Stability and Injury

The stability of the spine is an area that researchers explore to better understand the mechanisms of injury as well as the keys to a more efficient system. Reeves, Narendra and Cholewicki explored spinal stability and the standard well defined terms used to discuss spine dynamics, stability and injury. Stable behavior of the spine is critical to sustain load, allow movement, avoid injury and pain (Reeves, Narendra et al. 2007). They explored core and trunk stabilization literature and found that the concept of rehabilitation using core stability became more common. The core stability concept often implied more spine stiffness, however, they noted that the stiffer spine is not always the ideal environment when in situations that are more dynamic and precise rather than static. They went further to differentiate spinal stability and spinal robustness. The system is either stable or not and core stabilizing exercises do not make the spine more stable, they make it more robust, thus reducing risk of injury (Reeves, Narendra et al. 2007). They noted that increased muscle activation and co-activation (creating greater stiffness) does not mean more robustness always, since it is context dependent and sometimes a more compliant-faster responding system is better in a dynamic environment. Overall, co-activation creating more stiffness can mean a more stable spine in a static environment but in unstable environments or dynamic situations the goal should be to find the optimal level of trunk stiffness to ensure the task can be completed.

Neutral spine positions have been popular in back research and some researchers have argued that maintaining a neutral spine in exercise or daily life movements is the position of least loading for the spine (Cholewicki, Panjabi et al. 1997; Kiefer, Shirazi-Adl et al. 1997; Kiefer, Shirazi-Adl et al. 1998; Granata and Wilson 2001). It has been advocated as a safe place to begin exercise as well as a position of power and balance, however, athletic performance and functional activities move through this neutral zone (Panjabi 2003; Akuthota and Nadler 2004). Inclination of the vertebrae (and other ranges beyond a neutral spine) is an area that has not been explored. Previous studies have explored displacement of vertebrae to predict Nm of force that the spine is enduring but have not used accelerometrty to detect changes in lumbar inclination during fatigue. Accelerometry has been explored using high frequencies to study motor control (Ryan, Cramer et al. 2008), however, the lower frequencies can be used to detect movement and inclination (Webber and Kriellaars 2004). Using the lower frequencies can be helpful in exploring changes not only to muscle firing as they fatigue but also joint positions due to fatigue.

Measuring Trunk Stability during Movement

In the past, many studies have used electromyography (EMG) to gain insight into mechanisms of underlying trunk stability(Stevens, Vleeming et al. 2007) (Anderson and Behm 2005) (Arokoski, Valta et al. 2004) (Barnett and Gilleard 2005) (Cosio-Lima, Reynolds et al. 2003) (Davey, Lisle et al. 2002) (Hubley-Kozey and Vezina 2002) (Mori 2004) (Ng, Parnianpour et al. 2001) (Ng, Richardson et al. 2002) (Souza, Baker et al. 2001) (Vera-Garcia, Grenier et al. 2000) (Vezina and Hubley-Kozey 2000) (Walsworth 2004). However, there is little compelling information regarding activation strategy (motor control error) and its contribution to lumbar stability. Work has begun to focus on changes in motor control aspects of the lumbar spine from both reflex and volitional components (Zedka and Prochazka 1997). Direct measures of lumbar stability or motor control have not been established, yet accelerometry is developing as a tool which can provide a trunk stability measure. Accelerometry can provide more information about the mechanisms of fatigue including movement and inclination that EMG cannot.

Accelerometry has been used to provide a measure of motor recruitment (activation strategy) via mechanomyogram (MMG) (Esposito, Orizio et al. 1998) (Orizio, Diemont et al. 1999) (Ryan, Cramer et al. 2008). Mechanomyogram is a method of measuring muscle contractile behavior by measuring the vibration induced by muscle activation. Mechanomyogram is a method of measuring movement and recordings of muscle contractions by utilizing the high frequency component (> 4 Hz) of the acceleration data. Accelerometry has been used to measure movement related acceleration in the lumbar spine during upper and lower limb motion (Webber and Kriellaars 2004) and thoracic and lumbar accelerations in different body types during different movements (Ng, Bussone et al. 2006). MMG may provide more information regarding the interactions between motor unit recruitment and firing rate that control muscle force production than surface EMG (Ryan, Cramer et al. 2008). Using accelerometry as a measure provides the potential to explore three possible aspects of lumbar stability: activation strategy or motor control (high frequency), movement control (intermediate frequency) and lumbar inclination components (low frequency < 0.2 Hz).

Various researchers have explored trunk stabilization on unstable surfaces using EMG. In 2005 Anderson and Behm used EMG to examine trunk muscle activity in three different conditions: squats, smith squats and squats while standing on two balance discs (Anderson and Behm 2005). They found that there was a 33.8% ($p<0.01$) decrease in upper lumbar erector spinae EMG activity during the smith squat compared to the unstable squat (on two balance discs) and a 22.9% ($p<0.05$) decrease in the squats. They found that as the movements became more unstable there was only a negligible increase in the activity of the prime movers whereas there was a significant increase in the trunk stabilizer EMG activity. Behm and Leonard also explored trunk muscle EMG activity with unstable exercises in 2005 (Behm, Leonard et al. 2005). They used 11 subjects and compared shoulder and chest exercises on an unstable surface (swiss ball) or on a bench while performed bilaterally and unilaterally. They found that instability generated 27.9% greater lower abdominal activation during the chest press and 37.7%-54.3% increase in all trunk stabilizers. They felt that trunk strengthening should involve unstable surfaces.

In addition to the EMG activity it would be useful to know what happened with the trunk kinematics as well. Accelerometry could be an excellent tool in these scenarios to also measure the level of lumbar joint control, movement or inclination in the various unstable environments.

Motor control strategies are underrepresented in trunk stabilization exercise research, yet some studies have provided insight into the role of reflex mechanisms invoked in maintaining back motion control during limb motion. In 1997 Zedka and Prochazka explored phasic activity of the erector spinae during repetitive hand movements (Zedka and Prochazka 1997). They found that opposing side erector spinae EMG bursts were entrained to rapid hand movements. The subjects were not able to voluntarily contract the erector spinae in the same manner. During imposed rhythmical hand movements the erector spinae again remained entrained although the movement frequencies exceeded what was attainable by either the hand or back. They concluded that the hand movements elicited proprioceptive reflexes in erector spinae muscles that presumably contributed to trunk stabilization. Accelerometry could be used in a study such as this to examine joint movement or inclination during limb motion. Whether limb motion is entrained to lumbar control in an unstable bridge position is also unknown. The kinematics during the entrained contractions could provide more insight to the motor control strategies.

EMG has been used to explore ratios of trunk muscle contractions during the ball bridging (also referred to as ball plank) exercises and these exercises have also been used to compare activation of trunk stabilizers in normal subjects and back pain subjects(Stevens, Bouche et al. 2006). In 2006 Stevens et al. explored trunk muscle activity in healthy subjects during bridging stabilization exercises and found that in healthy subjects the back stabilization musculature contributes to control spine positions as does the abdominal musculature(Stevens, Bouche et al. 2006). They wanted to use ratios to investigate both relative (as a percentage of maximal voluntary isometric contraction) muscle activity levels and ratios of local to global muscle activity, during bridging stabilization exercises. They found different ratios of contribution of the abdominals depending on the bridge position. They found that the relative muscle activity and the ratio of the abdominal obliques seem to alter depending on the task. The ratios they found for the back muscles support the assumption that during these bridging exercises all back muscles contribute in a similar way to control spine positions and movements in a healthy population. The contribution of abdominal or back muscles or the kinematics of muscle fatigue in the bridging position has not been explored in a highly trained population.

In 2007 Schellenberg et al. studied lumbar spine endurance using prone and supine bridge maneuvers. Test-retest reliability measures of prone and supine bridging duration were recorded from 43 subjects without back pain and were compared with those of 32 subjects with chronic mechanical low-back pain (Stevens, Bouche et al. 2006). They found that supine bridge positions activated more trunk extensor muscles whereas the prone bridge positions activated more trunk flexors but both were compromised in subjects with back pain. Mean bridge durations for subjects without back pain were 72.5 +/- 32.6 (mean +/- SD) secs in prone and 170.4 +/- 42.5 secs in supine. They were significantly less in subjects with back pain: 28.3 +/- 26.8 secs in prone and 76.7 +/- 48.9 secs in supine. They found ball bridging exercises to be a reliable method to assess trunk musculature endurance and activation capabilities in healthy and low back pain patients. They did not explore the kinematics of failure and also did not examine a trained population.

Mechanomyogram research is relatively new yet a seemingly useful tool for assessing inclination and motor control. Esposito et al. explored electromyogram and mechanomyogram changes in fresh and fatigued muscle during sustained contraction in men (Esposito, Orizio et al. 1998). They aimed to examine the fatigue induced changes in electrical and mechanical properties of motor units. They tested the biceps brachii during a sustained 80% maximal voluntary contraction once before and 10 minutes after the fatiguing exercise. They found that the EMG values increased over the two tests and the MMG values decreased from the first test to the second and declined continuously during test two. They confirmed that both the mechanical and electrical activities of motor units were altered. Watakabe et al. explored the reliability of the mechanomyogram detected with an accelerometer during voluntary contractions in 2003 (Watakabe, Mita et al. 2003). They found that the MMG signal from the accelerometer accurately reflected the acceleration of body surface vibration during voluntary contractions of the quadriceps muscles. Accelerometry has been reliably used as a tool to detect changes in contractile properties of muscle.

Trunk Stabilization Training

Trunk stabilization training has become a major trend in rehabilitation, injury prevention, and athlete conditioning. Terms such as lumbar spinal stabilization, core strengthening, and postural stability training also describe trunk stabilization training (Akuthota and Nadler 2004). It is commonly accepted by therapists, coaches and trainers that some level of trunk stabilization ability is required to function in daily living tasks and excel in sporting movements (Barr, Griggs et al. 2005). Different programs and tools have been utilized to develop greater trunk stability such as stability balls and pressure transducers (Behm, Leonard et al. 2005) (Barnett and Gilleard 2005). Commercial tools to develop trunk stabilization have blossomed and include products such as stability balls, balance discs, Bosu apparatus, abdominal roll out devices, foam rollers and many more. Despite a large number of products there is not a variety of trunk stabilization measuring tools, techniques or an in depth understanding of what is being prevented during fatigue when trying to stabilize.

In 2004 Webber and Kriellaars studied the effect of stabilization instruction on lumbar acceleration (Webber and Kriellaars 2004). They positioned accelerometers over L1 and L4 spinous processes to record anterior/posterior acceleration during 10 repetitions of lower and upper limb movements before and after stabilization instruction. They found that each repetition exhibited four phases of acceleration. The instruction resulted in a significant reduction in peak lumbar acceleration magnitude (hip flexion 39.5% and biceps curl 30.8%) ($p<0.01$). They concluded that instruction could reduce lumbar acceleration levels due to a change in motor control strategy of trunk musculature. However, although they examined motion dependent accelerations, they did not use the accelerometers to examine activation strategy changes as a result of instruction, nor did they measure the absolute positional changes of the spine during the task. These two areas require further research.

Summary and General Statement of the Problem

There is a paucity of information regarding direct measures of lumbar stability. Short term adaptations in lumbar kinematics have been demonstrated as a result of lumbar stabilization training, largely instruction based and not exercise based (Webber and Kriellaars 2004). Ball bridging and trunk muscle activation or fatigue has not been explored using mechanomyogram. Trunk muscle failure or fatigue during a ball plank exercise has not been explored using high performance athletes. Furthermore, techniques for measuring trunk stability such as accelerometry could provide a greater understanding of the kinematics of the spine than EMG and further research into the use of accelerometry in back stability is required. Despite recent research using EMG to study trunk stabilization, no study to date has assessed lumbar stability using accelerometry to assess fatigue in trained athletes. The present study aims to describe the kinematics in trunk stabilization during a fatiguing bridging task on an unstable surface in highly trained athletes using lumbar accelerometry (mechanomyogram, inclination and differential acceleration). This will be done by examining inclination components, activation strategies, fatigue effects and differences in lumbar accelerations from person to person in a group of trained athletes.

Specific Objective of Research and Hypotheses

The main objective of the research was to measure lumbar kinematics during a fatiguing task (prone ball plank or bridge with elbow extension to move the ball for repetitions) in order to evaluate mechanism of failure in trained subjects

Hypotheses

1) There will be a change in lumbar kinematics as the trial progresses due to fatigue of trunk stabilization muscles:

   a) L1 inclination dependent upon repetitions

This would occur because of fatigue of the rectus abdominus, external obliques, transverse abdominus muscles, which would be contracting to prevent spinal extension to maintain the plank position. If these muscles fail in the prone position lumbar lordosis would occur causing an inclination change in L1 position.

   b) L4 inclination dependent upon repetitions

As with hypothesis a) this would occur because of fatigue of the rectus abdominus, external obliques, transverse abdominus muscles yet could be less than L1 due to the close proximity to the pelvis since it is a larger boney structure that has additional stabilizers.

   c) Lumbar orientation will change with repetitions

This would occur since it is a trial to failure and the subject may be orchestrating the recruitment of various trunk stabilizers to continue the task as long as possible. In doing so the lumbar orientation may change due to some abdominal muscles and spinal stabilizers fatiguing prior to others.

2) L1 and L4 inclinations will be correlated

Due to the demanding nature of the task the subject will show change in both lumbar vertebrae and they will be related.

3) Lumbar MMG magnitude will increase with repetitions consistent with fatigue.

4) L1 and L4 MMG will be correlated due to the co-activation of the trunk muscles surrounding the L1 L4 region.

Methodology

Design

The design was within group repeated measures.

Subjects

The subjects (n=10) were core stabilization trained athletes. A questionnaire (based on self report) was used to determine whether or not the subjects qualified as being core stabilization trained. Due to the high performance training the subjects regularly performed they qualified as core stabilization trained. Any subjects who were untrained and were utilized in the experiment were not included in the analysis since the trained athletes were used to analyze fatigue characteristics.

Inclusion Criteria

Subjects were between the ages of 19-31 years, males and females, Canadian National Team volleyball athletes and able to comprehend written and verbal English.

Exclusion Criteria

Anyone with a history of low back injury within the last year, a musculoskeletal injury that prevented them from exercise or the inability to comply with instructions were excluded from the study. Pregnant or lactating females were not used in this study.

Recruitment and Consent

The subjects were recruited by word of mouth and formed a sample of convenience. All subjects were required to provide consent prior to participation in this study. This study was ethically approved by the Education Nursing Research Ethics Board.

Protocol

The following is the timeline that the subject followed.

1. Informed consent
2. Subjects height (cm) and body mass (kg) were recorded
3. Questionnaires regarding level of training were filled out

4. Subjects were read a scripted verbal description and were shown a demonstration of the exercise
5. Range of motion markers were set
6. Accelerometers were affixed to the spinous processes of L1 and L4
7. Electrogoniometer was affixed to the subject's right arm
8. Back stability perturbation index (ball drop) was performed
9. Three sets of five repetitions of the ball moving plank task were performed with two minutes rest in between.
10. Back stability perturbation index (ball drop) was performed
11. Repetitions of a core stabilization exercise (ball moving plank) to volitional fatigue/ failure was performed
12. Back stability perturbation index (ball drop) was performed

For the purpose of this study, only the set to volitional fatigue/failure was processed and analyzed.

Exercise Description

The exercise (ball moving plank) is a trunk stabilization movement that is performed on a stability ball (Figure 1). The subject placed their feet on the floor and their elbows and lower arms on the ball. They contracted their trunk stabilization muscles and kept their body straight from their shoulders to their feet. The subject's upper arm should be in a starting position that is approximately perpendicular to the floor. The subject flexed their shoulder to move the ball forward approximately 5 cm and extended at the shoulder to move it back to the starting position.

Electrogoniometer Repetition Detection

A potentiometer (Bourns, 5 kOhm, 0.25% linearity) was affixed to the right upper limb so that the axis of rotation of the potentiometer was aligned with the elbow joint approximated by the lateral epicondyle. A stable voltage supply (10V) was applied to the potentiometer so that a voltage output was proportional to elbow joint angle. The potentiometer was calibrated using a two point, right angle calibration method. The voltage output was converted to elbow joint angle using the calibration coefficients derived from the two point calibration.

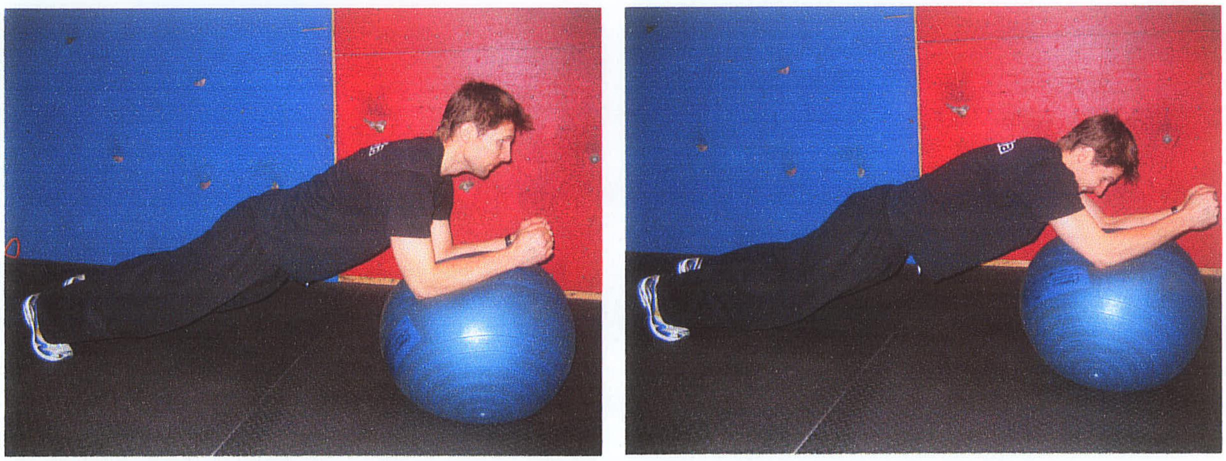

Figure 1 Ball Moving Plank Exercise-permission given by subject for use of photo (see appendix C)

Lumbar Accelerometry

A uniaxial accelerometer (EGAX series, Entran, USA) was connected to a bridge amplifier (PS30, Entran, USA) where the gain and offset was adjusted for maximal dynamic range and sensitivity (maximum gain with offset near zero voltage). The accelerometer was calibrated using gravitational orientation method (Webber and Kriellaars 2004). The accelerometer was placed on the skin overlaying the spinous processes of L1 and L4 vertebral bodies and oriented so that the sensitive axis in the anterior/posterior direction relative to the spine. L1 was land-marked by palpating the 12th ribs and following them medially and posteriorly to the spinous process of the T12 vertebral body. L1 spinous process was then be palpated one spinous process below the T12 spinous process. The L4 spinous process was then located by counting three spinous processes below L1. The output of the accelerometer was inputted to a data acquisition system and oscilloscope.

The recorded accelerometer data was processed by separating the signal into three frequency ranges (low, intermediate and high) corresponding to inclination of the accelerometer, movement acceleration, and activation strategy (tremor). The inclination corresponds to low frequency of the accelerometer signal from 0 Hz to 0.2 Hz. The movement based acceleration is found in frequency from 0.2 Hz to 2 Hz. The activation strategy (high) is found in frequency above 4 Hz.

The lumbar acceleration measures at L1 and L4 allow the assessment of four features through extraction of three different frequency components (items 1 and 2 below) and the derivation of differential acceleration. Rational for separation into these three components is illustrated in methodology section.

1. The low frequency component of lumbar acceleration provides a measure of lumbar inclination or postural changes during exercise.
2. The high frequency component that corresponds to the mechanomyogram of the surrounding working muscles and provides information related to motor unit activation changes.
3. The difference between L1 and L4 in intermediate and low frequency acceleration values provides a measure of lumbar motion segment control or orientation.

The variables to be derived from the lumbar acceleration data will include the magnitudes of

1. the high frequency acceleration of L1 and L4 respectively and
2. the low frequency acceleration of L1 and L4 respectively.

In addition the difference in acceleration between L1 and L4 in the intermediate and low frequencies will be measured to find any differences in sequencing of lumbar motion.

Data Acquisition and Sampling

A 16 bit analog to digital (A/D) converter (USB 9800 Data Translation, USA) was used to sample the accelerometer signals using an input range of +- 10V. The sampling rate was 1000 Hz (this will provide over-sampling for all of the signals) based upon the maximum frequency content of each of the signals determined by a power spectrum analysis (4096 point FFT, Sigview 1.95, USA). The majority of the signal is well below 30 Hz for accelerometers and below 10 Hz for the goniometer. Shannon's sampling Theorem states that a sampling rate that is 2x the maximum frequency of a signal is adequate to preserve the signal's original frequency. A rate of 1000 Hz exceeds this requirement.

Data Analysis and Processing

An example of raw data recorded during a trial is shown in Figure 2. In this figure, the person starts in an upright position (accelerometers oriented almost vertically), then adopts a position as shown in Figure 1. This change in orientation of the trunk is reflected in the low frequency, upward shift in baseline of the L1 and L4 accelerometer data (see Figure 2). During the ball plank task, the trunk is to be stabilized during arm motions. Changes in angular inclination of the L1 or L4 spine are reflected in baseline shifts of the L1 and L4 accelerometer data. At the end of the fatigue trial, the subject returns to a standing position. The electrogoniometer data is shown in Figure 2C. This data reflects the change in angle of the elbow with successive repetitions during the ball plank task. In this task, a repetition consists of a range of motion controlled concurrent shoulder and elbow motion. The initial motion is shoulder flexion with elbow extension followed by a return to the original position. An upward deflection in the elbow electrogoniometer data indicates elbow extension. Each repetition is automatically determined by software using a threshold algorithm (explained below). A marker for each repetition is shown with a vertical line (Figure 2C).

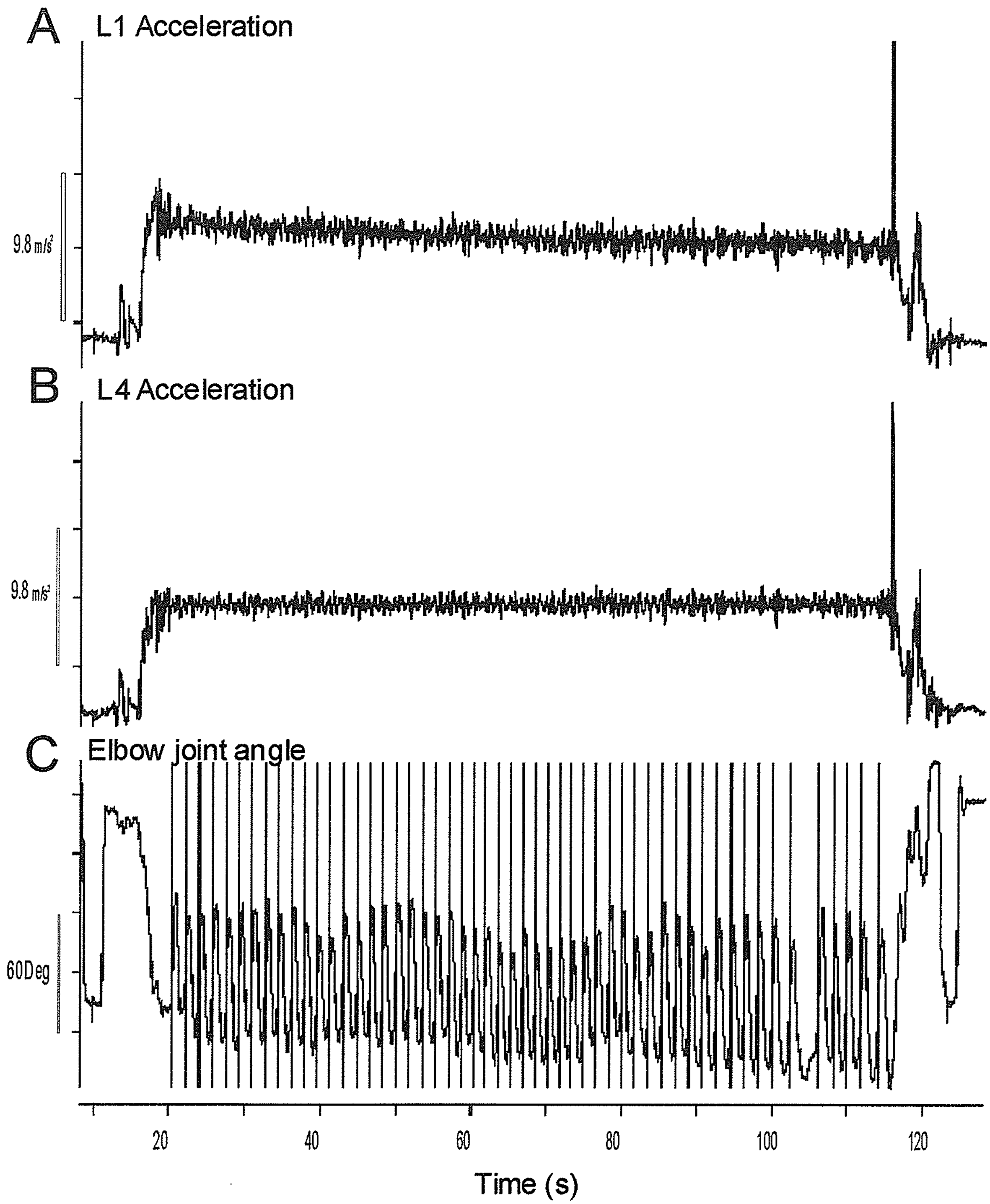

Figure 2 Measurements recorded during a single fatigue trials from a representative subject. A- L1 anterior/posterior acceleration B- L4 anterior/posterior acceleration and C elbow joint angle derived from electrogoniometry.

Figure 3 shows a brief portion of the fatigue trial. Panels A, B and C are the raw data waveforms. Selective signal processing was used to extract the desired components from the raw acceleration data (See Figure3). Panels D and E represent the mechanomyogram of the L1 and L4 accelerometer waveforms derived from panels A and B respectively. These MMGs were derived by high pass filtering (4 Hz) the accelerometer waveform at a frequency which removed the inclination and movement related acceleration frequencies. Thus these signals contain the high frequency data of acceleration which convey the muscular vibration information related to the neural control of the muscles surrounding the L1 and L4 regions. The MMG signals were further processed to obtain the linear envelope of the MMG data (panel K and L). A linear envelope entails taking the absolute value of the MMG waveform followed by performing a low pass filter set at a sufficiently low frequency to follow the contours of the waveform (much like a running average). This allows quantification of the magnitude of the L1 and L4 MMG data. Waveform C is the elbow joint angle raw data from the electrogoniometer. Waveform H is the low pass (0.2Hz) filtered electrogoniometer data so that any baseline shifts are removed to allow a single threshold to be used to demarcate the start of a repetition. Waveform I is band pass filtered (0.2 to 3Hz) for calculation of angular velocity seen in waveform J. Waveform J is elbow joint angle velocity mathematically derived from waveform I. Waveforms O and P are waveforms used to assess the change in inclination of the L1 and L4 spine, as they are low pass filtered versions of the raw acceleration waveforms (A and B). Waveforms O and P are also used to assess the lumbar orientation (the difference in inclination between L1 and L4. Waveforms F, G, M and N are the motion dependent waveforms consisting of band-passed frequencies of the raw acceleration data. This band pass removes the inclination information and the MMG from the signal – leaving the accelerations arising from the motion of the spine. Although processed from the original waveforms, the motion dependent accelerations were not analyzed or used in the results.

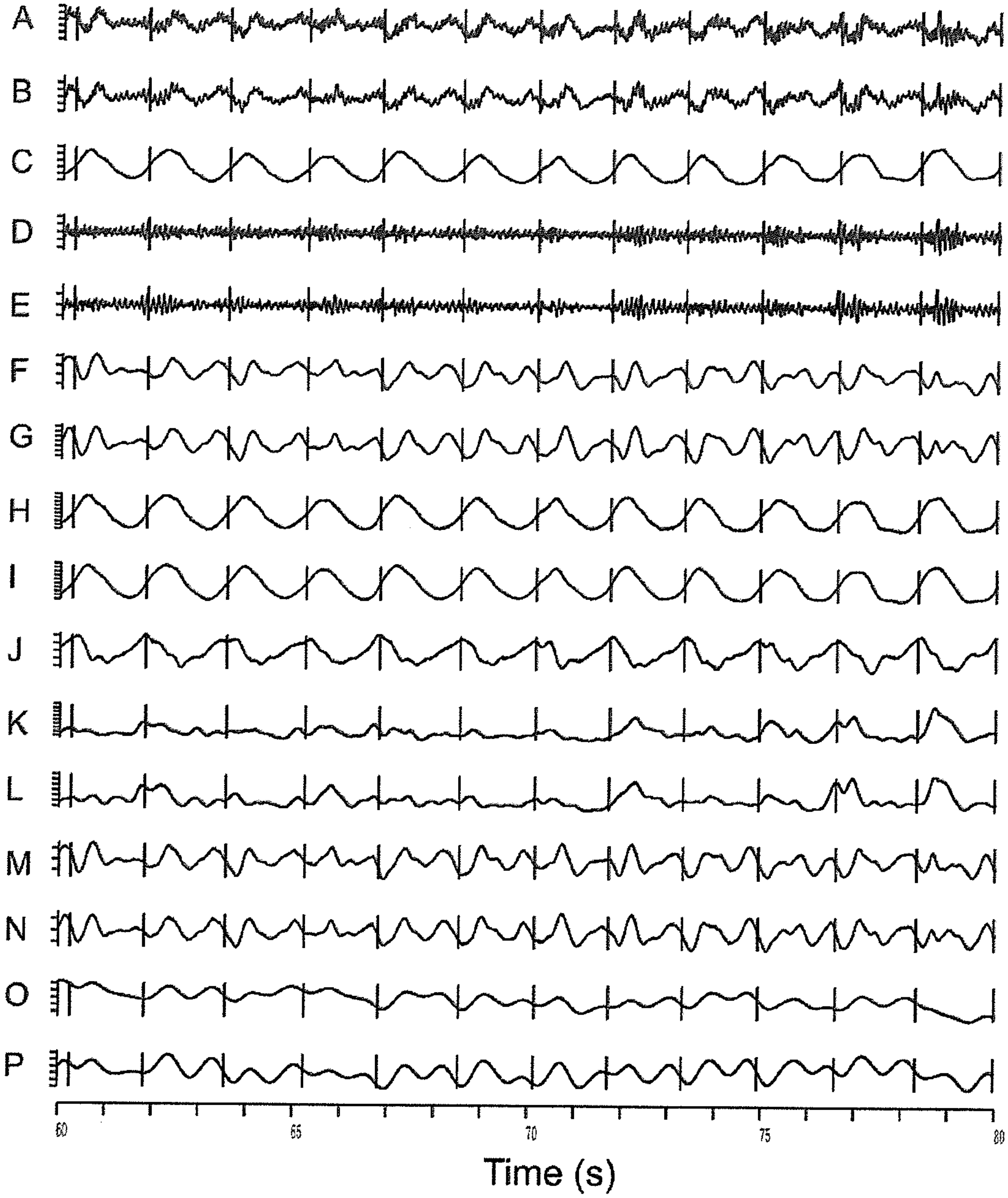

Figure 3 Raw and processed waveforms. A- L1 acceleration (unfiltered) B- L4 acceleration (unfiltered) C- elbow joint angle (unfiltered) D – L1 MMG (L1 acceleration >4Hz HP filtered) E- L1 MMG (L1 acceleration >4Hz HP filtered) F- Movement related acceleration at L1 (L1 acceleration < 3Hz LP filtered) G - Movement related acceleration at L4 (L4 acceleration < 3Hz LP filtered) – H AC coupled (> 0.2 Hz HP filtered) elbow joint angle data I – elbow joint angle data necessary (band-pass filtered 0.2- 3 Hz) for calculation of angular velocity (shown in J) J- elbow joint angle velocity (mathematically differentiated from I) K – linear envelope of L1 MMG (full wave rectified and then low pass filtered at 2 Hz) L - linear envelope of L4 MMG (full wave rectified and then low pass filtered at 2 Hz) M – band-pass filtered (0.2 to 3 Hz) of movement related L1 acceleration N – band-pass filtered (0.2 to 3 Hz) of movement related L4 acceleration O – L1 inclination (LP filtered L1 acceleration at 0.75Hz) P – L4 inclination (LP filtered L4 acceleration at 0.75Hz)

Statistical Analysis

Initially, the data were summarized and presented in the form of descriptive statistics including mean, standard deviation, range (max and min) and coefficient of variation when indicated. To examine the relationship between exercise parameters and lumbar kinematics, correlation, regression analysis and ANOVA were performed. The alpha level was set to 0.05.

Note that the test of significance for the slope gives exactly the same value of P as the test of significance for the correlation coefficient. Although the two tests are derived differently, they are algebraically equivalent.

Results

Subjects

The subject's physical characteristics are reported in Table 1. The subjects were all members of Canada's National Volleyball Team. Since they are high performance athletes they are accustomed to training regularly and maintaining a healthy body weight and a lean body composition. The subjects were healthy, fit, and did not report any recent back injuries. Since the subjects were National level volleyball athletes, the average height is taller than normal. Two of the athletes were female subjects with the remainder being male.

Table 1. Physical characteristics of subjects.

	Mean	SD	Minimum	Maximum
Age (yr)	24.8	3.12	21	30
Height (cm)	190.2	11.26	163	202
Mass (kg)	85.4	11.60	62	98.7
BMI	23.5	1.76	20.1	26.2

Exercise Parameters associated with fatigue

The subject's were instructed to do the exercise until failure, the time to failure and the repetitions to failure varied greatly (see Table 2). The repetition duration also had a large range which indicates that each subject found the rhythm that was best for them within the constraints of the range of motion markers, as represented by a low mean coefficient of variation of the repetition duration.

Table 2. Exercise parameters for the fatigue trial.

	Mean	SD	Minimum	Maximum
Repetition (#)	54.4	36.82	20	148
Time to failure (s)	72.8	31.12	42.8	144.9
Repetition duration (ms)	1601.0	540.3	875.9	2577.0
Coefficient of variation of repetition duration (%)	9.21	6.48	4.24	24.18

Correlations between Exercise Parameters

The inter-relationship between exercise parameters was examined by using correlation. There was a statistically significant and strong correlation between the number of repetitions and the time to failure ($r = 0.856$, $p=0.002$). The greater the number of repetitions, the greater the time to failure or the greater the set duration. This relationship is shown in Figure 4. In addition there was a significant, moderately strong inverse relationship between repetitions and repetition duration ($r = -0.689$, $p=0.028$). This relationship is shown in Figure 5. A relationship between repetition duration and time to failure was not observed ($r=0.28$, NS).

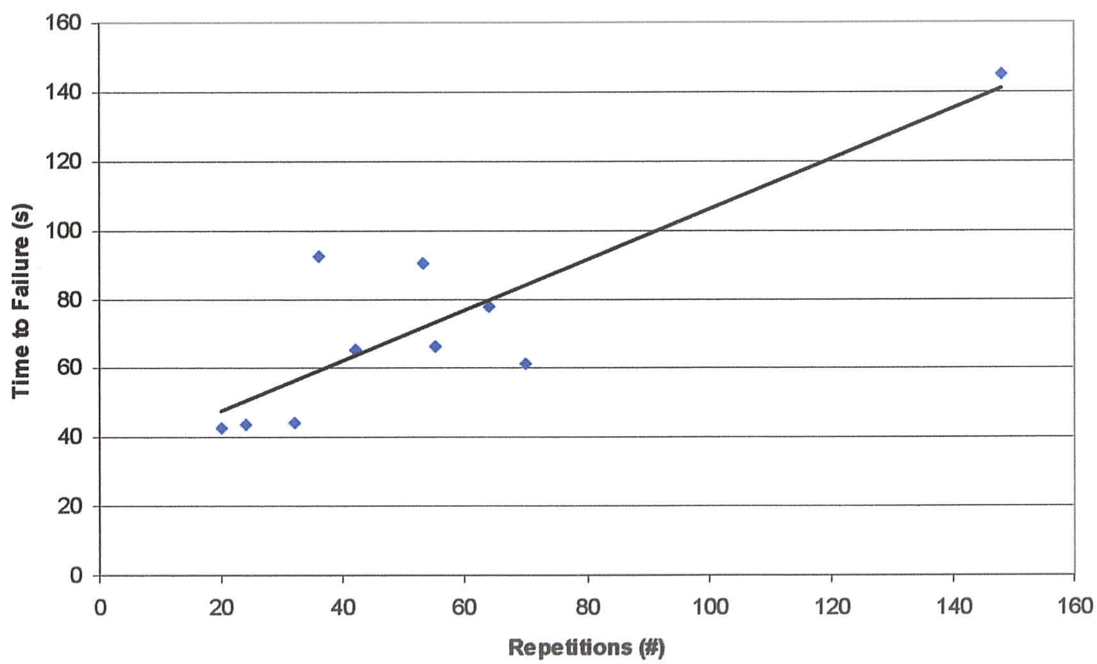

Figure 4 Relationship between time to failure (s) and number of repetitions. The greater the number of repetitions to failure, the greater the time to failure. Equation of line of best fit is $y = 0.7277x + 33.23$, $r= 0.856$, $p=0.002$.

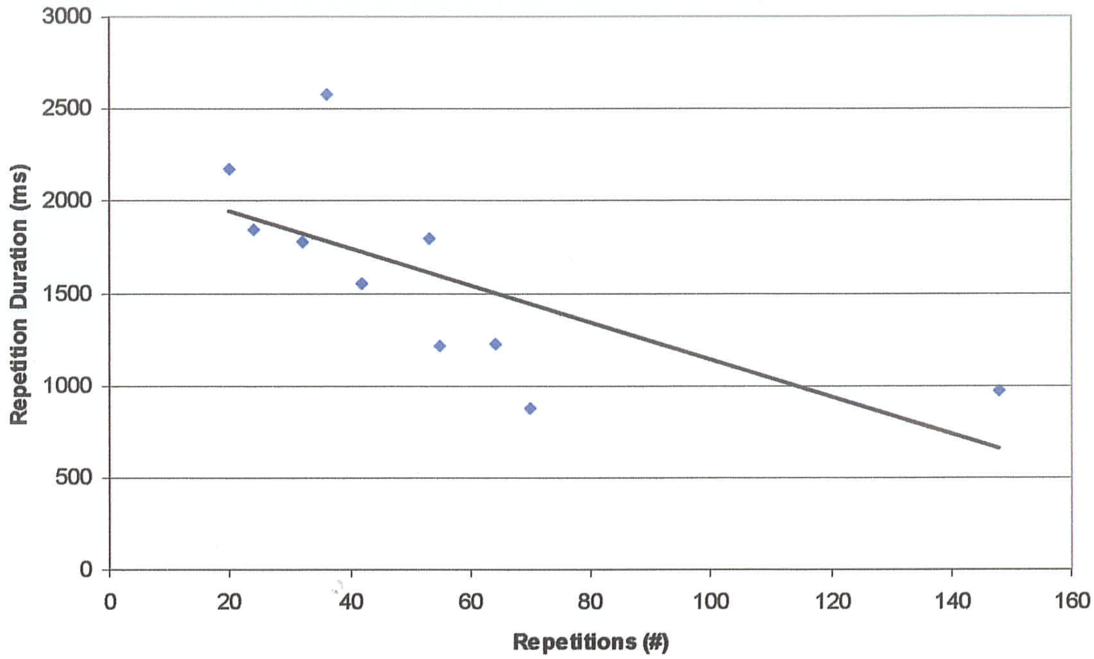

Figure 5 Relationship between the repetition duration (ms) and the number of repetitions. The greater the repetition duration, the fewer repetitions performed. The equation of the best fit line was $y = -10.054x + 2148$, $r = -0.689$, $p = 0.028$.

Relationship of exercise parameters and body size

The relationships between body parameters and exercise characteristics were examined and results are shown in Table 3. There was a negative correlation ($r = -0.663$, $p = 0.037$) between subject height and repetition duration (Figure 6) which also could be due to the horizontal position of the exercise since there is greater leverage in the taller people. When examining the relationship between body mass and repetition duration there is a strong negative correlation ($r = -0.749$, $p = 0.013$, Figure 7). The task difficulty is dependent upon the span of the bridge (plank) -therefore related to height and mass. A strong positive correlation ($r = 0.756$, $p = 0.011$)was present between sex and repetition duration indicating that the female athletes did slower repetitions as compared to their male counterparts, and this could represent the fact that females were shorter and less massive. Number of repetitions and time to failure were not correlated with height, mass, BMI or sex.

Table 3 Exercise parameters and body size

		HT	MASS	BMI	SEX
Repetition duration	r	-.663	-.749	-.366	.756
	p	.037	.013	.299	.011
Repetition	r	.310	.338	.154	-.378
	p	.384	.339	.671	.282
Time to failure	r	.134	.012	-.167	-.088
	p	.713	.974	.645	.808

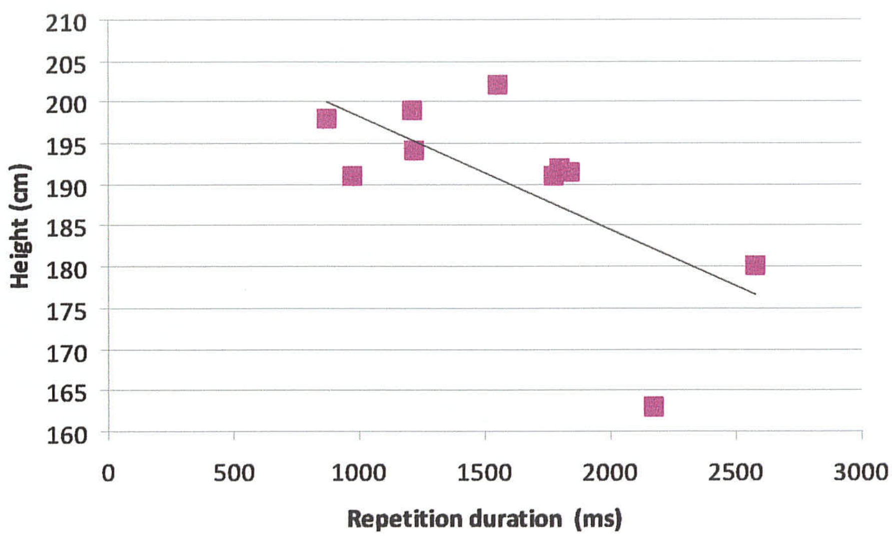

Figure 6 Relationship between repetition duration (ms) and subject height (cm). The taller the athlete, the faster the repetition. The equation of the best fit line was y = -0.0138x + 212.27, r= -0.663, p = 0.037

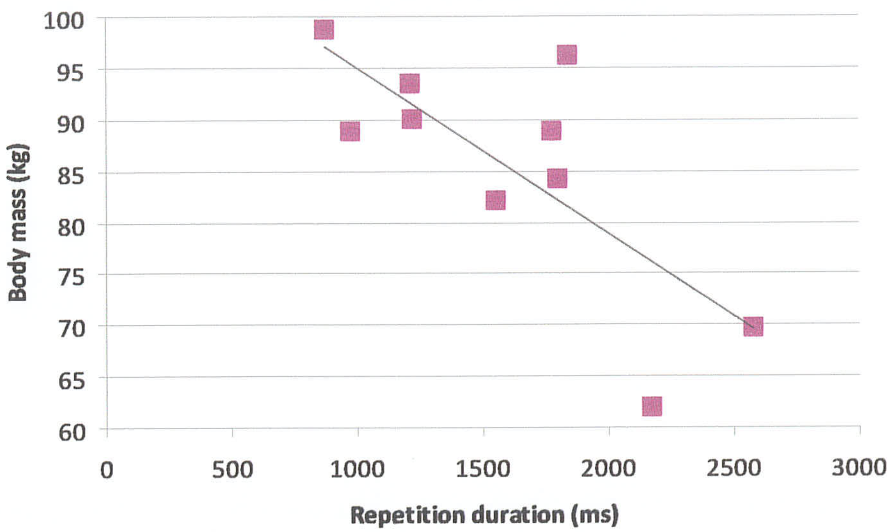

Figure 7 Relationship between repetition duration (ms) and body mass (kg). The greater the mass of a subject, the faster the repetition. The equation of the best fit line was $y = -0.0161x + 111.1$, $r=-0.749$, $p=0.013$.

Lumbar Inclination & Orientation

Lumbar inclination refers to the change in angular position in the L1 and L4 vertebrae as represented by the accelerometers affixed onto the spinous processes. Figure 9 shows the raw L1 and L4 accelerometer data (A and B) from a representative subject. Some subjects performed fewer or greater repetitions, however, this subject's data allows the viewer to see the repetitions more clearly versus how small the figure needs to be for a subject who performed greater repetitions. This subject represents what was typical of the majority of subjects. In order to observe inclination changes (Panels C and D), the raw signals were processed by low pass filtering (<0.2Hz) processed from A and B. Note the change in orientation in L1 inclination (Panel C) as the set to failure progresses. The inclination changes as the subject gets closer to failure whereas D (L4) stays relatively the same throughout the set. Figure 10 illustrates the magnitude of change over the set to failure (arrows). For each repetition the software automatically extracted the lumbar inclinations from the low pass filtered waveforms to produce inclination versus repetitions graphs shown in Figure 9. Linear regression was performed for each subject using this data (shown in Figure 9) to determine if significant changes in lumbar inclination occurred. The overall lumbar orientation, that is the difference in L1 and L4 inclination, was computed for each subject. Figure 10 graphically depicts the difference between L1 and L4 indicating the change that is occurring over the set to failure. Linear regression was performed between lumbar orientation and repetitions for each subject.

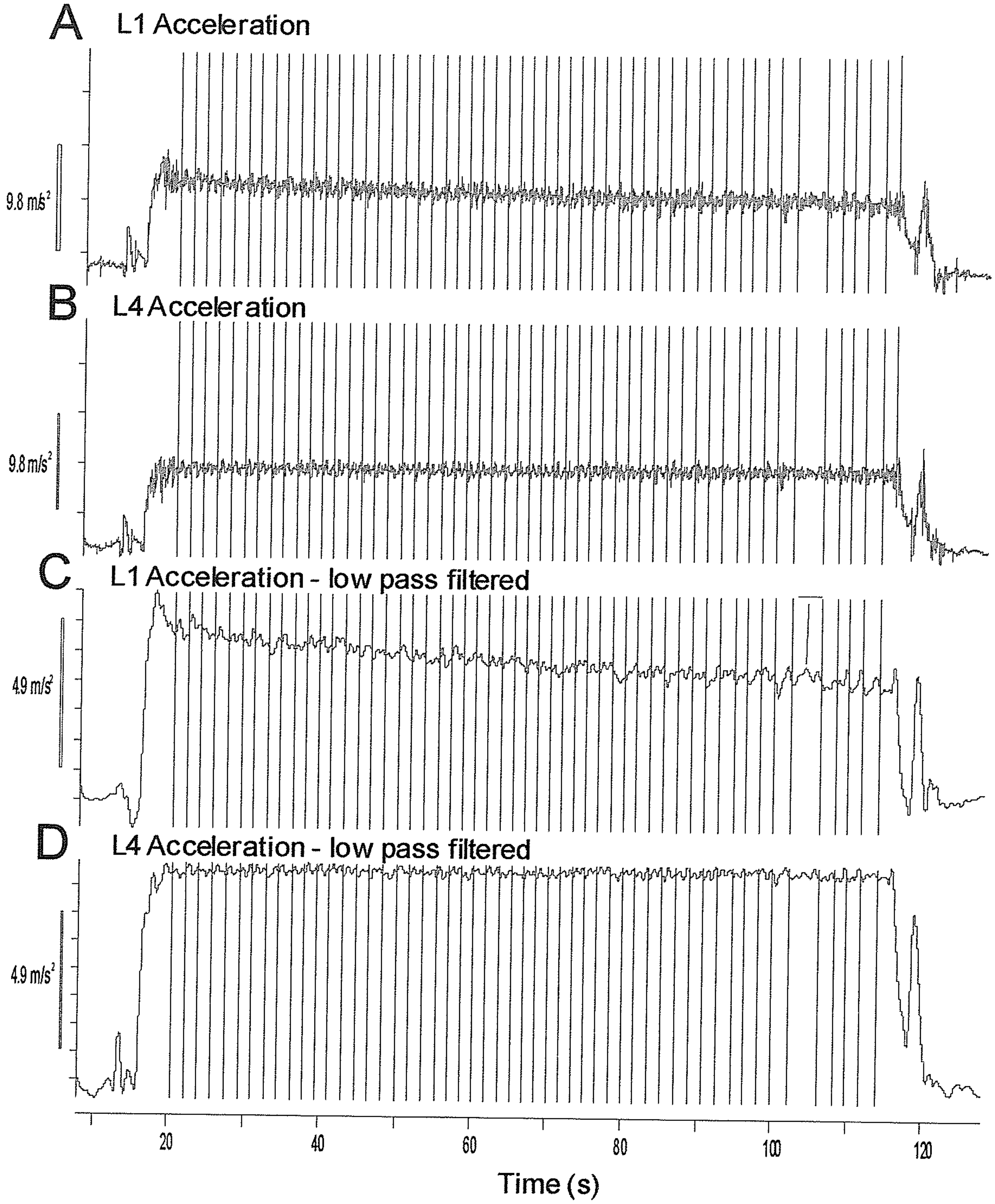

Figure 8 Low frequency acceleration data in a single representative subject. A-L1 raw acceleration data. B- L4 raw acceleration data. C- L1 inclination (LP filtered L1 acceleration data at <0.02Hz). D- L4 inclination (LP filtered L1 acceleration data at <0.02Hz). The vertical lines represent repetition markers which are indicated on the elbow joint angle data. The fatigue trials commences at the start of the first repetition marker.

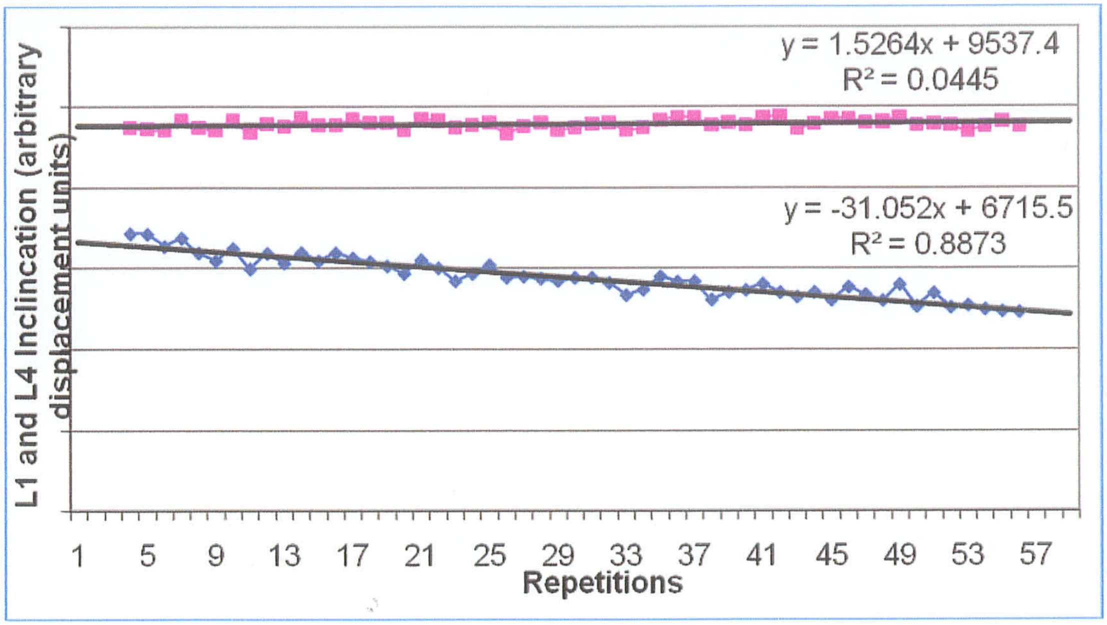

Figure 9 L1 L4 inclination changes over repetition- L1 is represented by the diamond shape and L4 represented by the square.

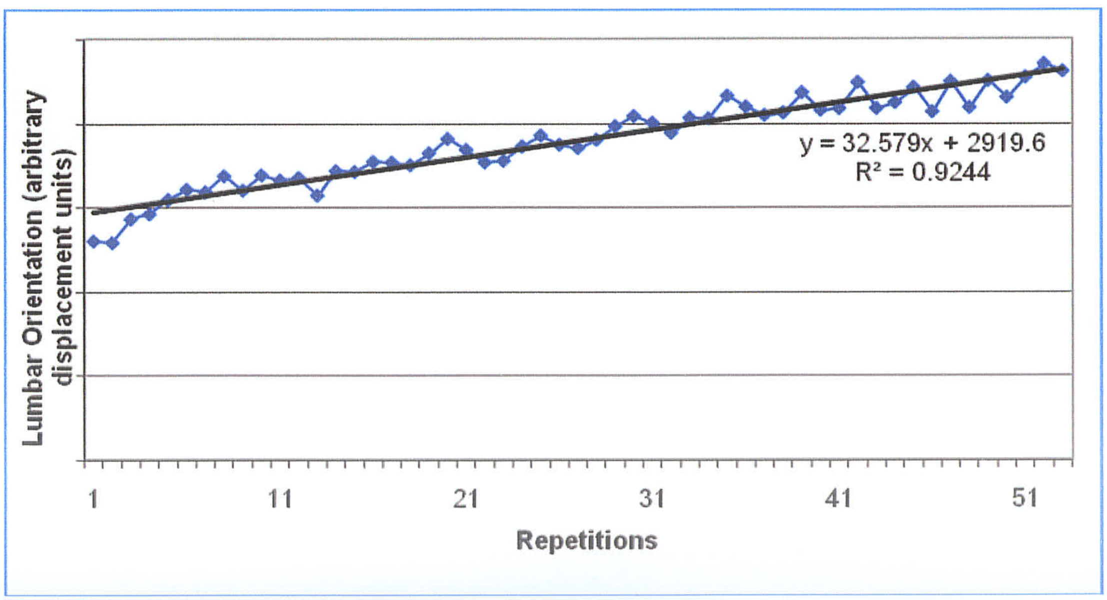

Figure 10 L1 L4 orientation. This is the difference between L1 and L4 occurring over the set to failure.

Table 4 shows the relationship between lumbar inclination/orientation and repetitions, as well as, the relationship between one segments inclination and overall lumbar orientation and the interrelationship between the motion at one segment and the other segment. Seven of 10 subjects revealed a significant relationship between L1 inclination and repetitions. Six of ten subjects revealed a significant relationship between L4 inclination and repetitions. Six of ten subjects revealed a significant relationship between L1-L4 orientation and repetitions. Eight of ten subjects showed a significant relationship between L1 and L1L4 orientation. Six of ten subjects showed a significant relationship between L4 and L1L4 orientation. Seven of ten subjects revealed a significant relationship between L1 inclination and L4. It is interesting to note that any subject that did not have a significant relationship between L1 and L1L4 did have a strong significant relationship between L4 and L1L4. The converse is also true. In reference to the interrelationship in motion between L1 and L4, one subject had a strong correlation (0.922) whereas three had moderate correlation and two had weak correlation and three had non-significant relationships. This partial independence in motion in L1 and L4 is illustrated in Figure 9 and 10 in one subject.

Table 4 Correlation between L1 and L4 inclination and repetitions. The correlation of orientation and repetitions, and L1 and L4 inclination and orientation is also shown.

	L1 - Reps	L4 - Reps	L1L4 - Reps	L1 – L1L4	L4 – L1L4	L1 to L4
1	0.124 NS	0.211 NS	0.148 NS	0.497 (<0.007)	0.989 (<0.001)	0.477 (<0.001)
2	0.477 (<0.001)	0.947 (<0.001)	0.839 (<0.001)	0.042 NS	0.864 (<0.001)	0.539 (<0.001)
3	0.518 0.001	0.494 (<0.001)	0.058 NS	-0.415 (<0.007)	0.642 (<0.001)	0.430 (<0.005)
4	-0.942 (<0.001)	0.211 NS	0.961 (<0.001)	-0.977 (<0.001)	0.232 0.094	0.019 NS
5	0.218 NS	-0.198 NS	-0.476 (<0.019)	-0.768 (<0.001)	-0.063 NS	0.688 (<0.001)
6	-0.549 (<0.003)	0.189 NS	0.567 (<0.002)	0.9 (<0.001)	0.468 (<0.014)	-0.037 NS
7	0.162 NS	0.930 (<0.001)	0.336 NS	-0.862 (<0.001)	0.136 NS	0.384 (<0.094)
8	-0.810 (<0.001)	-0.949 (<0.001)	0.146 NS	-0.674 (<0.001)	0.078 NS	0.684 (<0.001)
9	0.861 (<0.001)	0.924 (<0.001)	0.396 (<0.001)	0.071 NS	0.452 (<0.001)	0.922 (<0.001)
10	0.424 (<0.001)	0.9 (<0.001)	0.947 (<0.001)	0.334 (<0.005)	0.898 (<0.001)	0.715 (<0.001)

Analysis of variance (ANOVA) was used to analyze the group for inclination and orientation. ANOVA failed to detect a significant change in L1 with repetitions however, ANOVA revealed a significant change in L4 with repetitions ($p<0.001$) as well as a significant change in L1 L4 orientation with repetitions ($p<0.001$).

Individual changes in Lumbar Orientation

The L1 and L4 data for each subject was visibly examined to produce the following description:
Subject 1 showed a small change between L1 and L4 indicative of a lordotic curve. Subject 2 showed a change between L1 and L4, indicative of a lordotic curve. Subject 3 showed no change between L1 and L4 and did not indicate a curvature change over the course of the fatigue trial. Subject 4 showed a clear change between L1 and L4. A clear lordotic curve occurred over the course of the fatigue trial. Subject 5 showed a small change between L1 and L4 indicating a kyphotic curve as a result of fatigue. Subject 6 showed a change between L1 and L4, indicating a lordotic curve. Subject 7 showed a change between L1 and L4 but was a result of a shift toward kyphosis and then lordosis over the fatigue trial. Subject 8 showed a difference between L1 and L4 which represented a shift toward lordosis then a flattened position and then kyphosis.
Subject 9 showed a change between L1 and L4, indicative of a lordotic curve.
Subject 10 showed a clear change between L1 and L4, indicating a lordotic curve.
Therefore, six of ten subjects showed a clear change toward a lordotic curve, with one subject showing no clear change and two others showing kyphosis with a shift toward lordosis at the end of the trials. Eight subjects showed lordosis shift near failure.

Mechanomyograms

Representative mechanomyograms and associated derived signals are shown in Figure 13. The MMGs for L1 and L4 are shown in Panels C and D of Figure 13 which were derived from the raw lumbar acceleration waveforms (Panels A and B). In order to evaluate changes in magnitude of the MMG signal, the MMG waveforms were further processed to produce linear envelopes shown in Panels E and F. A linear envelope is produced by taking the absolute value of the MMG waveform followed by low pass filtering to obtain a signal which follows the amplitude trends of the MMG waveforms. In the bottom panel of Figure 13, the phasic nature of the lumbar MMG is evident and synchronized to the repetitions performed about the elbow.

For a representative subject, Figure 14 depicts the peak magnitude of the linear envelope MMG for each repetition at L1 and L4 respectively. For this subject, the magnitude increases as the fatigue trial continues to volitional failure, although demonstrating some variability. This trend toward increasing MMG amplitude for both L1 and L4 is typical for most subjects and was further analyzed using correlation.

For each subject, correlation was performed between the linear envelope MMG amplitude and repetitions (See Table 5). The mean correlation coefficient between L1 and repetitions was 0.52 with nine of ten subjects showing significant positive correlations. The mean correlation coefficient between L4 and repetitions was 0.51 with eight of ten subjects showing significant positive correlations, and one subject at the verge for significance ($p=0.057$).

The interrelationship between L1 and L4 MMG magnitudes was examined with correlation (Table 5). All subjects demonstrated positive correlations between L1 and L4 MMG but with a large range in the correlation coefficients ($r=0.44$ to 0.93). The mean correlation coefficient between L1 and L4 was 0.76.

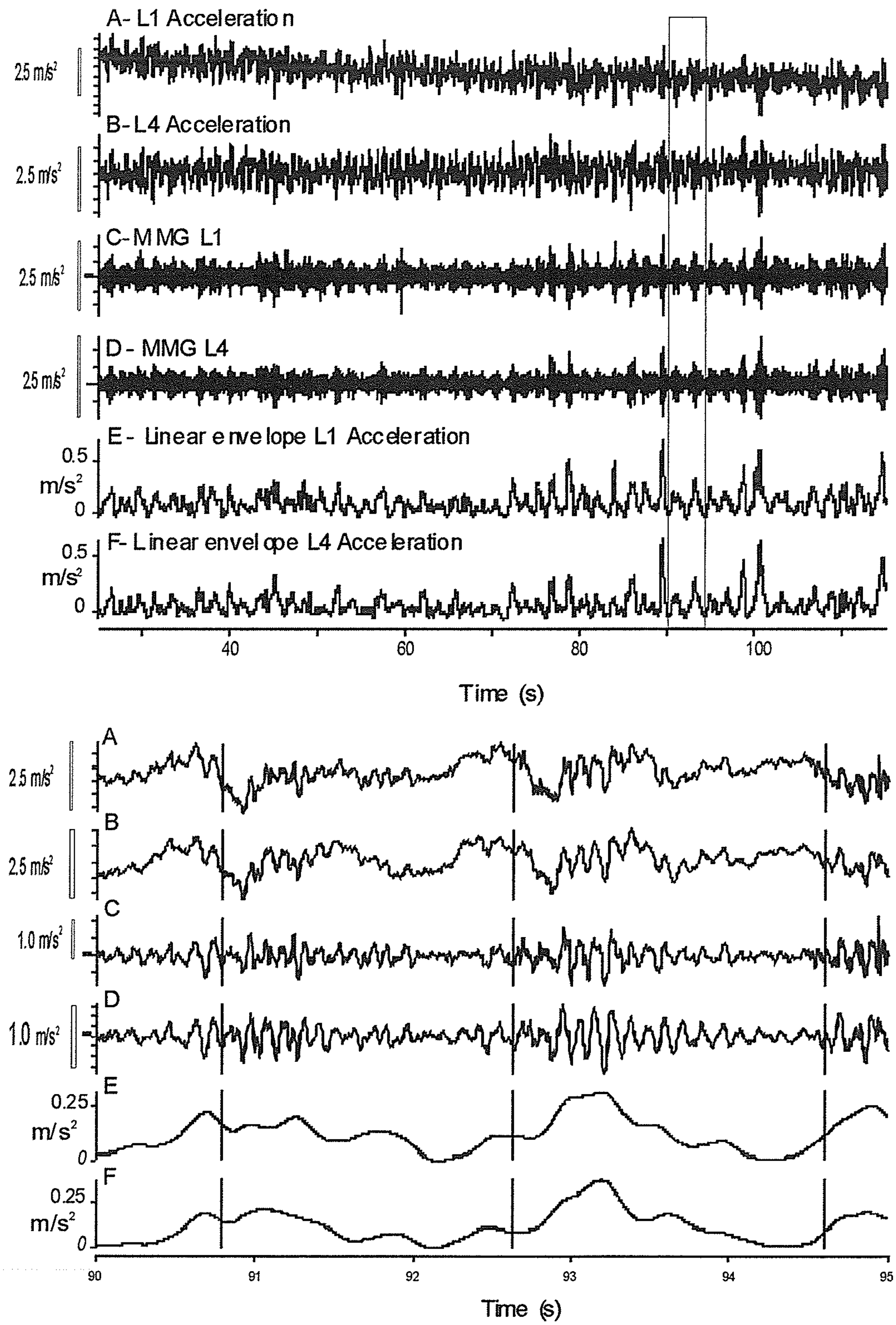

Figure 11 Mechanomyograms for a representative subject (Top graph- a full fatigue trial, Bottom Graph – a close up of the signals). Top: A and B are the L1 and L4 raw acceleration waveforms. C and D are the MMG waveforms for L1 and L4. E and F are the linear envelope waveforms corresponding to L1 and L4. Same designation for the Bottom graph.

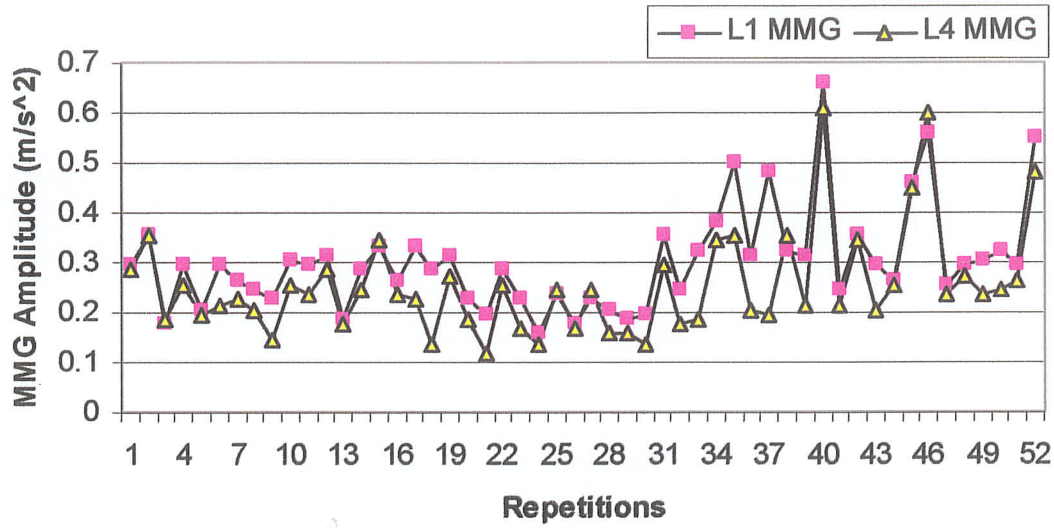

Figure 12 The relationship between MMG magnitude and repetitions for a representative subject. The magnitude of the MMG was automatically extracted per repetition by the software and depicted graphically. L1 MMG magnitude is shown in diamonds while L4 magnitude is shown in squares. A significant and substantial increase in MMG amplitude is observed over repetitions.

Table 5 Correlation between L1 & L4 magnitude and repetitions (r and p values). In addition, the relationships between L1 and L4 are shown (all significant, p< 0.001).

	L1 vs repetitions		L4 vs repetitions		L1 vs L4
Subject	r	p value	r	p value	r
1	0.62	p < 0.001	0.37	p< 0.01	0.60
2	0.59	p < 0.001	0.16	NS	0.44
3	0.61	p < 0.001	0.74	p < 0.001	0.75
4	0.41	p< 0.01	0.34	p< 0.01	0.87
5	0.55	p< 0.01	0.63	p < 0.001	0.93
6	0.47	p< 0.01	0.45	p< 0.01	0.84
7	0.36	NS	0.43	p=0.057	0.89
8	0.34	p<0.05	0.57	p < 0.001	0.51
9	0.75	p < 0.001	0.74	p < 0.001	0.89
10	0.48	p < 0.001	0.65	p < 0.001	0.88
Mean	0.52		0.51		0.76
SD	0.13		0.19		0.18

ANOVA was used to analyze the group for significant differences. ANOVA revealed a significant ($p=0.01$) change in L1 MMG magnitude with repetitions across all subjects. ANOVA also revealed a significant ($p<0.001$) change in L4 MMG magnitude with repetitions. Regressions between L1 and L4 MMG magnitudes across all subjects revealed a significant positive relationship ($r= 0.936$, $p<0.001$) between the two variables.

Discussion

Summary of Results

Results demonstrated that as subjects fatigue in an unstable ball plank task they tend to perform repetitions at a rate that is unique to their body size and gender. The subjects who were taller did faster repetitions as did those who were male. If the mass of the subject was greater the repetitions were faster. The subjects that did the repetitions slower performed fewer repetitions to failure. There was not a significant relationship between L1 inclination and repetitions but there was between L4 inclination and repetitions ($p<0.001$). ANOVA revealed a significant relationship between L1 L4 orientation and repetitions ($p<0.001$). Eight of ten subjects showed a lordotic shift near failure with one showing no clear change and one showing a kyphosis. The interrelationship between L1 and L4 MMG magnitudes was examined with correlation. All subjects demonstrated positive correlations between L1 and L4 MMG but with a large range in the correlation coefficients ($r=0.44$ to 0.93). The mean correlation coefficient between L1 and L4 was 0.76. ANOVA revealed a significant ($p=0.01$) relationship between L1 MMG magnitude and repetitions across all subjects. ANOVA also revealed a significant ($p<0.001$) relationship between L4 MMG magnitude and repetitions. Regressions between L1 and L4 MMG magnitudes across all subjects revealed a significant positive relationship ($r=0.936$, $p<0.001$) between the two variables.

L1 L4 Inclination and Orientation

Trunk stabilization fatigue may influence neuromuscular control of the spine, thereby affecting the inclination of the vertebrae under load and potentially increasing the likelihood of injury. The robustness of the spinal stabilizers is put to the test in the ball plank task and once the subjects fatigue the stability of the spine becomes compromised (Reeves, Narendra et al. 2007) leading to movement, in this case inclination, of the L1 vertebrae. Inclination becomes the change in position due to the horizontal position the subject is in during the task. The anatomical position of the lumbar spine is a lordodic curve and in a horizontal position the curve is magnified, especially during fatigue, without the strength of the abdominal musculature to stabilize the horizontal position. In this study there was not a significant relationship between L1 inclination and repetitions but there was between L4 inclination and repetitions ($p<0.001$). The subjects in this study are National level athletes who are trained to hold plank type positions with proper form. If their form was becoming compromised they would likely stop since they are trained not to compromise the quality of an exercise.

Furthermore, the subjects revealed a significant relationship between L4 inclination and repetitions but the correlations that were significant were stronger than that of the L1 inclination and repetitions correlations. Anatomically, L4 is situated closer to the pelvis and due to the abdominal (rectus abdominus, external obliques, transverse abdominus) attachments contracting against the tension of the psoas major (Akuthota and Nadler 2004) and the weight of the trunk, the abdominals are likely to fatigue and allow the anterior rotation of the pelvis which could influence the position of the L4 vertebrae. The anterior rotation of the pelvis is a greater movement, than a single vertebra, since it is a larger structure. That may cause the subject to be much more aware of its movement and terminate the task to failure if they sense that the pelvis is beginning to anteriorly rotate. In an untrained subject they may fail and simply allow the spinal passive structures to take the load which may put them at risk for injury or greater wear and tear on the vertebral joints.

The overall lumbar orientation is the difference in L1 and L4 inclination indicating the change in the L1 and L4 position in relation to each other. ANOVA revealed a significant relationship between L1 L4 orientation and repetitions ($p<0.001$). Although not all the subjects showed a significant relationship between L1 and L4 there were other relationships between L1 and the L1 L4 orientation and the L4 and L1 L4 orientation. Eight of ten subjects showed a significant relationship between L1 and L1L4 orientation. Six of ten subjects showed a significant relationship between L4 and L1L4 orientation. It is interesting to note that any subject that did not have a significant relationship between L1 and L1L4 did have a strong significant relationship between L4 and L1L4. The converse is also true. It would seem that L1 and L4 can be controlled separately but this is not likely something the subject is aware of or can control volitionally.

Interestingly, both female subjects did not have a significant relationship between L1L4 orientation and repetitions or L4 and L1L4 orientation, but did have a significant relationship between L1 and L1L4 orientation. None of the male subjects showed this same pattern of results. Upon reviewing the spine position of each subject, the two female subjects were two of only three that showed any kyphosis during the trial. One showed a difference between L1 and L4 but was a result of a shift toward kyphosis and then lordosis over the fatigue trial. The other showed a difference between L1 and L4 which represented a shift toward lordosis then a flattened position and then kyphosis. The fact that the female subjects did not have a significant relationship between L4 and L1 L4 orientation and that they showed kyphosis during the trial may indicate that they were attempting to stabilize their pelvis more so than their male counter parts. The remainder of the subjects all shifted toward a greater lordotic curve over the course of the trial.

The interrelationship in motion between L1 and L4, was such that one subject had a strong correlation (0.922) whereas three had moderate correlation and two had weak correlations and three had non-significant relationships. This partial independence in motion in L1 and L4 is illustrated in Figure 10 and 11 in one subject. Figures 10 and 11 are examples of a non-significant relationship. It would seem that the control of the spine during fatigue to failure can be independent between vertebrae but is dependent on the individual's muscular strength and endurance to maintain the stability (Reeves, Narendra et al. 2007). It is likely that the local trunk muscles fatigue and cannot maintain the spinal segments. Multifidi, transverse abdominus and internal obliques would be key in maintaining the segmental control (Akuthota and Nadler 2004) necessary in a ball plank task such as the one in this study.

The hypotheses tested that

1) there will be a change in lumbar kinematics as the trial progresses due to fatigue of trunk stabilization muscles:

a) L1 inclination dependent upon repetitions was not a significant relationship for the group as shown by the ANOVA findings. There was however, some significant relationships in the individual subjects.

b) L4 inclination dependent upon repetitions was a significant relationship for the group ($p<0.001$).

c) Lumbar orientation changed with repetitions since ANOVA revealed a significant relationship between L1 L4 orientation and repetitions ($p<0.001$)

2) L1 and L4 inclinations were correlated since the interrelationship in motion between L1 and L4, one subject had a strong correlation (0.922) whereas three had moderate correlation and two had weak correlation and three had non-significant relationships.

Implications of Inclination and Orientation findings

The concept of training trunk musculature (core training) in athletes is very popular due to the belief that the athletes can develop more power in sporting movements and prevent abdominal and back injuries with core training(Barr, Griggs et al. 2005). The purpose behind the core training is often to develop more stability throughout the spine. However, according to Reeves et al stability is either present or not-the goal rather should be to develop greater robustness(Reeves, Narendra et al. 2007). The concept of a more robust spine would allow the spine to endure stress and change in dynamic environments or static environments. In our subjects a more robust spinal system could endure the static contractions of the ball plank task with dynamic movements simultaneously. A more robust system can endure sporting movements and the demands placed on the system in various angles and unexpected perturbations in everyday life without sustaining injury.

A trained system that is more robust should be able to sustain static contractions on an unstable surface to fatigue while dynamic contractions are occurring, without injury. This study shows that multiple motor patterns are evident during fatigue as evidenced in table 4. Figure 10 also illustrates this, as it is clearly shown that L1 inclination may change while L4 remains the same and vice versa-as seen in our subjects. This type of segmental difference shows that some areas are maintaining stabilizing contractions while others are fatiguing and allowing movement. Kavcic et al found that no single muscle dominated in the enhancement of spine stability, and their individual roles were continuously changing across tasks. They felt that to train for stability and enhanced motor patterns it would be justifiable to incorporate many muscles rather than targeting just a few (Kavcic, Grenier et al. 2004). It has also been shown that subjects will co-contract with less force if the timing of the perturbation is known, leading to stabilization without as much compressive force (Vera-Garcia, Elvira et al. 2007). Targeting many muscles and multiple movement patterns is more sport and life specific due to the dynamic demands that sport and life place on many muscles and motor patterns. This can justify using exercises that may be considered contra-indicated for the general population but would be indicated for those who are in contra-indicated movements or positions that require spine stabilization (athletes, labourers, etc). It can also justify the use of perturbations in training to simulate certain unexpected jarring movements in sport and life.

We have demonstrated that during a core exercise to failure in highly trained athletes there is still spinal movement at the vertebrae angularly (inclination) and segmentally (orientation). This is important since these subjects (athletes who are accustomed to controlling their spine) were trying to stabilize and may have stopped the task upon perceiving such movement. Our findings have shown that during the ball plank task the vertebrae will not remain stationary although the subject is attempting to control any movement. This attempt to curb vertebral movement would lead to co-contraction and thus a more stable spine but according to some studies should not remain stiff (Reeves, Narendra et al. 2007). The spinal stiffness from co-contraction may create a system more likely to be stable but will also create more spinal compression (Vera-Garcia, Brown et al. 2006) (Lee, Rogers et al. 2006). It may be beneficial for the subject to maintain stiffness in a more static environment but a dynamic environment may require more variable dynamic control to maintain spinal stability. If the subject has the ability during a fatiguing task or perturbation to maintain or return to the approximate original vertebral position the system is stable or asymptotically stable according to Reeves et al.

If the disturbed behavior or "new behavior" differs significantly from the old behavior the system is unstable. It has also been found that conscious adjustments may be less safe than natural muscle activation pattern that would be present in a robust system (Brown, Vera-Garcia et al. 2006). This further confirms how complex the system is and how control of the spine is very context dependent.

In our study the task was unique in that there was a static component of stabilizing the spine while a dynamic movement at the elbow and shoulder joint was occurring. The greater leverage of the arm movement and the fatigue produced the "new behavior" or change in vertebral position, and the subject could correct it and maintain a stable environment or if they were unable to correct- further change in vertebral position would occur creating an unstable environment, potential injury and inevitably volitional failure. None of the subjects were injured but they stopped the task since they could no longer control their spinal stability. It would seem worthwhile to educate athletes and non athletes that training the spine in such a way that promotes absolute co- contraction and stiffness is not ideal. To train with more stiffness in static environments and less where more motor control is required is more likely to prevent injury and optimize performance. For instance, if our subjects exhibited too much stiffness in the ball plank task they would not have been balanced enough to merely maintain their position on the ball. There are however, situations that may require greater stiffness and compression on the spine such as a hockey body check (Reeves, Narendra et al. 2007), or lifting a heavy object.

If a more compliant and quick responding system is one that is robust in moving environments then it would make sense to train such a system in a manner that is specific to the needs of the athlete, laborer or patient. For instance, if core endurance is required in one position- then training in that position in static held contractions for time makes sense- especially if the individual has difficulty maintaining a contraction for any length of time. However, if the individual is in need of a responsive core system that needs to control the spine at multiple angles and rates of contraction and they are already able to contract voluntarily, then it would seem vital to train the spine in multiple angles and rates of contraction with a greater load or stimulus to create a more robust spine in that environment. This would be logical for non-athletes as well or for those who have suffered low back pain due to injury. Based on this study and previous literature, a series of trunk (core) training progressions that stimulate the nervous system to adapt and become more robust can be very useful for athletes, non-athletes and back pain patients and would be structured as follows:

1. Static contractions on a stable surface
2. Progressions (to more difficult exercises, angles, planes of movement and leverages)
3. Static contractions on an unstable surface (such as a ball, foam roller or balance board etc)
4. Progressions
5. Dynamic contractions on a stable surface
6. Progressions
7. Dynamic contractions on an unstable surface
8. Progressions
9. Dynamic contractions on a stable or unstable surface utilizing perturbations, ballistic movements, elastic resistance, manual resistance and sport and life specific complex patterns.
10. Progressions

Each new stage begins with the most basic exercises and leverages and can quickly progress to the next level if the individual is pain-free, and can display proper form and muscle activation at each level. A trained athlete could be at stage 1 if they are returning from injury whereas a construction worker may be at stage 9 based on the level of "training" they receive through daily work. An individual could also be able to train at 2 stages properly prepare them for the next level up. They may be at stage 4 and stage 5 to best prepare for the most basic level of stage 7. This can be a way that therapists and strength coaches can ensure that they have not

a) stayed too basic for too long and not properly prepared the individual for their sport or life thereby increasing their risk of injury when they return to their activities or

b) skipped key progressions and placed the individual in a stage that is too advanced for their strength or neurological capabilities and they use poor movement patterns to try and accomplish the task without the sufficient ability to do it properly thereby increasing their risk of injury when they return to their activities.

Some subjects, who were non athletes, were observed but not used for the data analysis. Of those subjects most of them allowed their back to shift into lumbar lordosis as they fatigued. This was an interesting observation since they continued the task although their abdominal muscles had fatigued and carried on until their limbs fatigued. They essentially maintained the plank position via the passive stiffness of their spine (boney and ligamentous structures) and perhaps did not perceive the change in position or could not regain any abdominal contraction or support. Untrained individuals such as these should initially be taught how to contract and maintain the contraction in the static environment before attempting more complex movement sequences. It would seem that if they cannot maintain a contraction in a still environment they may struggle with the contraction organization and sequencing required in dynamic environment.

L1 L4 Mechanomyogram

MMG has been shown to be a reliable tool for measuring the mechanical properties of muscle contraction and has not yet been utilized to its full potential (Watakabe, Mita et al. 2003). The high frequency of the accelerometer signals (MMG) allows us insight to the mechanisms of fatigue in this task. As the subjects fatigued during the ball plank task the lumbar MMG magnitude increased with repetitions. There was a significant relationship in the magnitude of the L1 MMG with repetitions in nine of ten subjects and a significant relationship in the magnitude of the L4 MMG with repetitions in nine of ten subjects This is consistent with EMG studies that show an increase in magnitude with muscle fatigue (Esposito, Ce et al. 2005). Our results showed that although there was a relationship with L1 and L4 and repetitions in nine of ten subjects not all of them showed an increase in magnitude over repetitions. Three subjects showed a decrease in magnitude of the MMG over repetitions while all of the others showed an increase.

The decrease in magnitude of the MMG for the three subjects could be due to a time incurred physiological muscle fatigue but this is unlikely considering almost all of the remaining subjects performed their set to failure in a greater amount of time. Another possibility is that there is a shared distribution of contractile responsibilities. As some muscles fatigue and they cannot produce the same contractile force and other muscles or passive tissues provide support. Recently, researchers have found that superficial back muscle fatigue seems to induce a shift in load sharing toward passive stabilizing structures (Descarreaux, Lafond et al. 2008). This was referred to as a flexion relaxation phenomenon. Typically no single muscle dominates in the enhancement of spine stability, and their individual roles are continuously changing (Kavcic, Grenier et al. 2004). Another possibility is that the MMG increased to a certain percentage of maximum voluntary contraction and then decreased. A study using MMG and rectus femoris fatigue found that the MMG increased progressively with force up to 70% MVC, beyond which it decreased (Akataki, Mita et al. 1999). Another possibility is that the subject may have "given up" trying to co-contract as much as the task became harder, thereby not activating the erector spinae muscles as much. The fact alone that the mechanomygram is showing changes over fatigue could be indicative of changes in recruitment since Lumbar inclination and orientation changes are occurring at the same time.

The hypotheses also examined:

3) Lumbar MMG magnitude for the group increased with repetitions consistent with fatigue. ANOVA revealed a significant ($p=0.01$) relationship between L1 MMG

magnitude and repetitions across all subjects. ANOVA also revealed a significant ($p<0.001$) relationship between L4 MMG magnitude and repetitions.

4) L1 and L4 MMG were correlated due to the co-activation of the trunk muscles surrounding the L1 L4 region. All subjects demonstrated positive correlations between L1 and L4 MMG but with a large range in the correlation coefficients ($r=0.44$ to 0.93). The mean correlation coefficient between L1 and L4 was 0.76.

Implications of Mechanomyogram findings

Overall, the mechanomyogram findings show that the ball plank task with shoulder flexion and elbow extension is fatiguing the erector spinae muscles. In a static plank position the abdominal muscles and the erector spinae muscles are co-contracting to maintain stability during the task. As the subjects fatigued the magnitude of the mechanomyogram was increasing in most subjects and the phasic magnitude increases were shown during the elbow extension phase. This type of task (static stabilizing with dynamic limb movement) could be a useful way to train since planned bursts of increased force of contractions could be a better orchestrated way to "perturb" the system in a statically contracted environment. For example, a volleyball player diving for a ball would have a sufficiently co-contracted trunk with an additional leverage and perturbation upon reach and contact with the ball or even more so with contact with the ground. Also, L1 to L4 MMG relationship were quite variable indicating that different segmental levels of muscle activation were utilized – when attempting to train a person using one form of activation strategies may be insufficient for all subjects – as they need to adapt "their' method of activation during task.

Subjects and Exercise Parameters

Results that were also interesting included the repetitions and time to failure. The greater the number of repetitions, the greater the time to failure- this was expected. However, what was unexpected was that the repetition duration did not correlate to the time to failure. Fast or slow repetitions did not dictate the time a person would fail in. A relationship between repetition duration and time to failure was not observed (r=0.28, NS). Another finding was that the repetition duration was correlated inversely to the number of repetitions performed. The greater the repetition duration, the fewer repetitions performed (figure 6). It would seem that the subjects had a strategy that was individual to their body since the repetition duration had a low coefficient of variation (mean 9.21%). There was a strong negative correlation between the repetition duration and the height indicated in Table 3. This could indicate that the position the subject is in for the exercise (prone and horizontal) causes height to be a significant factor since the leverage of a taller person is so much greater. There was also a strong negative correlation between the repetition and the mass (shown in table 3) which could be explained by the nature of the exercise since the subject must support their body mass while performing the exercise and may not have the strength to hold the repetition longer with a heavier body mass. Another correlation was between sex and repetition duration (see Table 3). The female athletes performed the repetitions slower than their male counterparts; this is expected as this would coincide with the height and mass relationships.

Implications for Exercise Parameter Findings

Considering that height and mass influence the strategy that a subject chooses to do a task to failure that involves leverage it may be useful to instruct subjects or athletes to do the task at the cadence that may be most useful for their sport or life movements. Considering the change in rate of movement we see with the taller and more massive subjects and the variable MMG relationship between L1 to L4, different activation strategies with exercise may be needed to find what is optimal for those who are taller and more massive. Individuals in an externally loaded state appear to select a natural muscular activation pattern appropriate to maintain spine stability sufficiently (Brown, Vera-Garcia et al. 2006). This is true based on the results of this study and is related to the gender, height and mass of the subjects.

Conclusions

In conclusion, a trunk stabilizing task to volitional failure in highly trained athletes leads to fatigue-related kinematic changes. The ball plank task with elbow extension caused phasic magnitude increases in the mechanomyogram of the erector spinae muscles attached the vertebrae to which the accelerometers were attached. The L1 and L4 inclination changed dependent upon repetitions and most subjects showed a lordotic shift near failure. Lumbar orientation also changed with repetitions and the subjects all displayed some level of movement in L1 or L4 vertebral bodies as they fatigued during the task. The inclination and orientation changes however, were such that they were related but moving differently. For example, L1 magnitude (figure 10) changed notably whereas L4 did not. This is consistent with other research that suggests that superficial back muscle fatigue seem to induce a shift in load-sharing with other muscles or passive structures to maintain stability. In this case it may have been the passive structures since the vertebral body moved. MMG magnitude increased with repetitions suggesting muscle fatigue. The L1 to L4 MMG relationship was quite variable indicating that different segmental levels of muscle activation were utilized. Training a person using one form of activation strategies may be insufficient for all subjects – as they need to adapt "their' method of activation during task. It was also noted that rate of movement was orchestrated dependent on mass and height. Our results suggest that given the way that fatigue-related kinematics occur in highly trained athletes it would be useful to customize trunk stabilization training to consider height, mass and cadence of movement. Furthermore, fatigue-related changes occur in multiple muscles inducing mechanical movement prior to volitional failure leading to various other muscles or passive structures to continue with the task- one must acknowledge that a complex orchestration of multiple contractions occurs to continue a task while other muscles are too fatigued to continue. This should be considered when training or rehabilitating individuals since fatigue, height and mass will change the kinematics of the task.

Recommendations for Future Studies

It would be useful for future studies to explore the following:

1. Using trained, untrained and back pain patients could be relevant to do a similar study to explore the fatigue-related changes between groups. This could allow insight to the kinematics of the lumbar spine, time to failure, and neurological properties during or after a trunk stabilizing task to failure. This could allow a greater understanding of appropriate rehabilitation or training progressions for each population.
2. Repeat the study using EMG and MMG with the MMG on joints and EMG on abdominal musculature to gather more information on what muscles are contracting as others fatigue. This could give researchers more understanding of what areas fatigue first in trained or untrained populations. This could allow more effective rehabilitation and training.
3. A study using trunk stabilization and limb movement such as this one, that controls for cadence. One group could have cadence controlled and the other chooses their own rate of movement. This could indicate whether time to failure and fatigue related kinematic changes are different based on height, mass or even trained or untrained subjects.

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Appendix A RESEARCH PARTICIPANT INFORMATION AND CONSENT FORM

Title of Study: Trunk Stability Measured with Lumbar Accelerometry and Electromyography: A Comparison of Trained and untrained Groups.

Principle Investigator: Kari Schneider, Rm. 138 Frank Kennedy Centre, 474-6645
Advisor: Dr. Philip Gardiner, Rm. 303 Max Bell Centre, 474-8770
Co-Investigator: Dr. Dean Kriellaars, RR303 Rehabilitation Hospital, 787-3505

This consent form, a copy of which will be left with you for your records and reference, is only part of the process of informed consent. It should give you the basic idea of what the research is about and what your participation will involve. If you would like more detail about something mentioned here, or information not included here, you should feel free to ask. Please take the time to read this carefully and to understand any accompanying information.

This will take approximately 15 minutes to complete.

Purpose of Study

The present study aims to describe the differences in trunk stabilization between trained and untrained groups. This will be done using accelerometry and electromyography. This will be done by examining movement components, muscle contractions, fatigue effects and differences in back accelerations between trained and untrained groups during a trunk stabilization exercise on an unstable surface (a ball).

A total of 24 subjects will participate in this study

Study Procedures

All participants are being recruited for this study via word of mouth and poster. You will be asked to attend one exercise session at the University of Manitoba Bannatyne Campus, Human Performance Laboratory, Rehabilitation Hospital, Health Sciences Centre, Winnipeg. Total time for this session will not exceed one hour.

Day of the week and time of day of this session will be scheduled to accommodate your schedule.

You will be asked to fill out a questionnaire to verify which group you will be placed in. At the exercise session, you will be asked to fill out a screening Par Q questionnaire to ensure you are able to exercise. Your body weight and height will also be measured.

Prior to performing any exercise you will view a demonstration and description of the Ball Plank exercise to be performed. Range of motion markers will be set and the lightweight miniature accelerometers (small device used to measure movement) will be taped to your lower back. The electromyogram (used to measure magnitude of muscle contraction) will be placed on the back muscles beside the accelerometers. A plastic electrogoniometer (used to detect range of motion) will be secured with tape to your forearm to measure the range of motion during the exercise. You will then perform an electromyography normalization task which consists of a maximal voluntary contraction (contract your back muscles as hard as you can): rest: submaximal contraction: rest. You will then be asked to catch a 1 kg ball in your outstretched hands (Ball Drop) and to perform three sets of five repetitions of the Ball Plank exercise with rest between each set (this will be demonstrated and described so you know how to do it). You will be asked to perform the Ball Drop again and subsequently a final set of the Ball Plank exercise to volitional fatigue or failure. You will once again be asked to perform the Ball Drop task. Input from the investigators may be provided to ensure safe and proper technique.

The exercise session will be stopped if:
-You wish to stop for any reason
-You exhibit signs of pain or severe discomfort
-You use unsafe technique in the performance of the exercise.

The researcher may decide to remove you from the study if you are unable to perform the exercise properly (i.e. safely).

You can stop participating at any time. However, if you decide to stop participating in this study, we encourage you to tell the study staff why you chose to stop.

Risks and Discomforts
After the exercise session, it is common to feel some soreness in the core muscles involved in performing the exercise. This very minor discomfort could last up to 5 days peaking at 2 days after the exercise. This is a normal consequence of exercise and is called delayed onset muscle soreness.

Minor discomfort may be felt during the exercise session (as with any form of exercise). However, if obvious pain arises at any time during the session, the session will be discontinued.

<u>Recording Devices Used</u>
Electromyogram will be used to record the muscle activity.
Accelerometer will be used to record movement.
Electrogoniometer will be used to measure the range of motion of the elbow.
These measurement devices pose no health risk.

Benefits

There may or may not be direct benefit to you from participating in this study. We hope the information learned from this study will benefit other people in the rehabilitative and resistance training settings.

If you wish to learn of the study results please feel free to contact Kari Schneider at 474-6645.

Costs

All the procedures, which will be performed as part of this study, are provided at no cost to you.

Payment

You will receive no payment or reimbursement for any expenses related to participating in this study.

Confidentiality

Information gathered in this research study may be published or presented in public forums, however your name and other identifying information will not be used or revealed.
Prudent measures will be taken to ensure that confidentiality is maintained.
Each subject will be assigned a study number and the subject's name will be deleted from the data. Kari Schneider, Dr. Dean Kriellaars and Dr. Philip Gardiner are the only people who will have access to the study data. The data will be kept in a locked lab on computers that are password protected and in file cabinets that are locked. The data will be kept for ten years (in a locked room and password protected) and will be disposed of at ten years via file deletion and paper shredding.

The University of Manitoba Education/Nursing Research Ethics Board may review records related to this study for quality assurance purposes.

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to participate as a subject. In no way does this waive your legal rights nor release the researchers, or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time, and/or refrain from answering any questions you prefer to omit, without prejudice or consequence. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation. If any questions arise during or after the study or if you have a research related injury, contact: Kari Schneider at 474-6645.

This research has been approved by the University of Manitoba Education/Nursing Research Ethics Board. If you have any concerns or complaints about this project you may contact any of the above-named persons or the Human Ethics Secretariat at 474-7122, or e-mail margaret_bowman@umanitoba.ca. **A copy of this consent form has been given to you to keep for your records and reference.**

Statement of Consent

I have read this consent form. I have had any questions regarding the study answered by the study staff in a language I understand. The risks and benefits have been explained to me. I understand that my participation in this study is voluntary and that I may choose to withdraw at ant time.

I understand that information regarding my personal identity will be kept confidential, but that confidentiality is not guaranteed. I authorize the inspection of any of my records that relate to this study by The University of Manitoba Research Ethics Board, for quality assurance purposes.

By signing this consent form I have not waived any of the legal rights I have as a participant in a research study.

Participant signature____________________________
Date____________________

Participant printed name____________________________

I, the undersigned, have fully explained the relevant details of this research study to the participant named above and believe that the participant has understood and has knowingly given consent.

Printed Name__________________________
Date_________________

Researcher and/or Delegate's signature____________________________________

Appendix B Training Questionnaire

1. Do you do core (trunk) strength and stabilization training? _____Yes _____ No

2. If yes, what type of training do you do?

   ____

   ____

   ____

3. Have you been training your core for six months or more? ______Yes ______No

4. Have you ever used a stability ball or other unstable surface for your training? _____Yes ______No

5. Do you regularly use unstable surfaces to train on or with? _____Yes _____No

6. Do you train with static (non-moving) contractions? _____Yes _____No

7. Do you train with dynamic (moving) contractions? _____Yes _____No

8. Are you employed at a job that requires heavy lifting or physical labor on a regular basis? _____Yes _____No

9. Do you train your core:______ 1x /week ______2x/week _____3x/week ______4x/week?

10. Have you endured a low back, hip flexor, or abdominal injury within the last year? ____Yes ____No

11. Have you endured a low back, hip flexor, or abdominal injury within your lifetime that still gives you pain and/or dysfunction? _____Yes _____No

APPEDNIX C Letter of Permission

July 15th, 2008

To Whom It May Concern,

I give permission for Kari Schneider to use my photograph as necessary. It may be used for her thesis or any future publications. I understand that it may be viewed in multiple scenarios and approve of this use.

Sincerely,

Chris Wolfenden