2014

THE STUDY OF TRUNK MECHANICAL AND NEUROMUSCULAR BEHAVIORS

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THE STUDY OF TRUNK MECHANICAL AND NEUROMUSCULAR BEHAVIORS

THESIS

A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in Biomedical Engineering in the College of Engineering at the University of Kentucky

By
Brian D. Koch
Lexington, Kentucky
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ABSTRACT OF THESIS

THE STUDY OF TRUNK MECHANICAL AND
NEUROMUSCULAR BEHAVIORS

Low back pain (LBP) is a common ailment in the United States, affecting up to 80% of adults at least once in their lifetime. Although 90% of LBP cases are considered nonspecific, recent studies show that abnormal mechanics of the lower back can be a major factor. One method of assessing the lower back mechanical environment is through perturbation experiments. An intensive literature review of perturbation systems was used to select and develop a system for the Human Musculoskeletal Biomechanics Lab (HMBL). Following construction, individuals with high/low exposure to day-long physical activity were assessed to quantify daily changes in their lower back mechanics and determine whether complete recovery occurs during overnight rest. Despite significant decrease in maximum voluntary contractions (MVC), intrinsic stiffness of the high exposure group remained constant following day-long physical activity. The final component of this Master's project is devoted to the design of a wobble chair system for study of trunk stability. Development of the perturbation system and wobble chair are hoped to facilitate future research aimed at a better understanding of trunk mechanical and neuromuscular behaviors to prevent and treat LBP in the future.

KEYWORDS: Low Back Pain, Perturbation Systems, Trunk Mechanics, Disturbance and Recovery, Wobble Chair

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July 29, 2014

THE STUDY OF TRUNK MECHANICAL AND
NEUROMUSCULAR BEHAVIORS

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I would like to thank Floyd Taylor for constructing the wobble chair and providing valuable insight on our design.

I would also like to thank my family, my friends, University of Kentucky's Ultimate Frisbee Team, and everyone in the Department of Biomedical Engineering. You have made graduate school fun and unforgettable. Thank you for your continued support.

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List of Abbreviations

AE:	After Exposure
AR:	After Recovery
BE:	Before Exposure
EMG:	Electromyography
FF:	Final Force
HMBL:	Human Musculoskeletal Biomechanics Lab at the University of Kentucky
HPA:	High Level of Physical Activity
IF:	Initial Force
K:	Lower Back Intrinsic Stiffness
L1, L2…L5:	Lumbar Vertebrae from Level 1 to 5
LBP:	Low Back Pain
LPA:	Low Level of Physical Activity
M:	Lower Back Apparent Mass
MVC:	Maximum Voluntary Contractions in any directions
MVE:	Maximum Voluntary Extension
MVF:	Maximum Voluntary Flexion
S1:	Level 1 of the Sacrum Vertebrae

Chapter 1: Introduction

Back pain is the second most common neurological ailment in the United States, being surpassed only by headaches (NINDS, 2004). It has been shown that ~ 70-80% of adults experience at least one episode of back pain in their lifetime (Rubin, 2007). Prevalence of back pain is higher in women than men and increases with age; ~ 22% of elderly people (ages 68-100) have been suggested to experience back pain symptoms "on most days" (Edmond & Felson, 2000). Back pain is also one of the most common symptoms prompting physician visits (Deyo & Phillips, 1996). Between 2% & 5% of the population has been estimated to seek medical attention annually due to back pain highlighting the significance of its associated cost (Rubin, 2007). In general, individuals with back pain spend 60% more on health care than individuals without back pain (Luo, Pietrobon, Sun, Liu, & Hey, 2004). Overall, in 1998 Americans spent around $90 billion on back pain related expenditures.

A majority of back pain is primarily located between the L1-L5 vertebrae and is referred to as low back pain (LBP). Most acute LBP cases generally last between a few days and a few weeks and are most likely the result of an identifiable trauma. LBP is considered chronic if it lasts beyond 3 months. Although it can result from untreated acute back pain, the cause of the progressive pain is often unknown (NINDS, 2004). Approximately 90% of LBP cases are considered to be nonspecific. The remaining 10% specific cases consist primarily of injuries and diseases (i.e., spinal fractures, degenerative disc disease, cancer) that can be identified from a radiograph and treated as necessary (Manek & MacGregor, 2005). There are, however, many known risk factors to be associated with LBP. These include demographic factors like age and gender, health factors like body mass index, occupational factors such as awkward working posture and lifting, psychological factors like stress, and abnormalities within the spinal anatomy (Rubin, 2007).

The spine is made of 24 vertebrae (excluding the sacrum's vertebrae) separated from each other by intervertebral discs that along with surrounding ligaments provide a high degree of flexibility and mobility to the trunk. Although the vertebral column provides flexibility, mobility, and attachment sites for trunk muscles, it is inherently unstable and requires assistance from active and passive trunk tissue to assure its stability. The synergy between the active and passive trunk tissues for both spinal equilibrium and stability will determine the state of lower back mechanics. Recent studies (Adams, 2004; Panjabi, 2006) have shown that abnormal mechanics of the lower back can lead to LBP. As such accurate assessment of mechanical behavior of the lower back is important for control and management of LBP. All trunk tissues can contribute to the equilibrium and stability requirements of the spine passively while select tissues (i.e., muscles) can also do so actively. The active contributions of muscles could be either voluntarily or reflexively and can be affected by both intrinsic (fatigue) and extrinsic (vibration, exercise) factors. The passive contribution of all trunk tissues is time dependent (viscoelastic) and is also affected by extrinsic factors. Although direct measurements of trunk tissue contribution to its mechanics are not possible, indirect measures can be used to estimate changes in tissue contribution to lower back mechanics. In particular stiffness, damping, and apparent mass of the lower back can be estimated using several different statics and dynamics experimental protocols to determine changes in tissue contribution to lower back mechanics. For instance, passive trunk deformations along with viscoelastic models can be used to identify changes in passive contribution of trunk tissues following exposure to a LBP risk factor or following a treatment for LBP. Perturbation experiments can be used to study alterations in active voluntary and reflexive muscle contribution in lower back mechanics.

The Human Musculoskeletal Biomechanics Lab (HMBL) in the Department of Biomedical Engineering is a newly developed lab directed by Dr. Babak Bazrgari at the University of Kentucky. One of the main research focuses of the lab is biomechanics of the lower back aiming at control and management of LBP. To equip the lab with the most accurate and reliable tools for assessment of lower back mechanics, this project was defined to develop and test a set of testing apparatuses to study the contribution of trunk tissues into the lower back mechanical environment. Following an extensive review of literature on the existing designs that have been used to evaluate lower back mechanics, two designs were adopted and were built for the HMBL: 1) a displacement controlled sudden perturbation system which can estimate changes in different aspects of trunk mechanics (i.e., passive, active voluntary, and active reflexive), 2) a wobble chair for empirical estimation of spine stability.

Organization of thesis:

In the following chapters, a review of existing sudden perturbation systems and protocols for evaluation of lower back mechanical behaviors is presented (Chapter 2). This review includes systems and protocols used to assess passive and active tissue contributions to the lower back mechanical environment. This review is then followed by a presentation of the design that was adopted and built. This is succeeded by an article (Chapter 3), presenting a small study that has been conducted using the system to study the effects of level of physical activity on active voluntary and passive lower back mechanical behaviors and their recovery. The article is followed by a brief overview (Chapter 4) on the improvements made to the perturbation system and experimental protocols based on the small study experience. Finally, a brief review of systems that have been used to evaluate seated spinal stability is presented (Chapter 5) followed by a presentation of the design adopted and built for HMBL.

CHAPTER 2: STUDY OF LOWER BACK MECHANICAL ENVIRONMENT USING SUDDEN PERTURBATION METHODS

2.1 Introduction

One technique of measuring the contributions of trunk tissues to its mechanical properties is through perturbation tests. Merriam-Webster describes a perturbation as a disturbance of motion, course, arrangement, or state of equilibrium (Merriam-Webster). Trunk perturbations happen daily, whether it is from something as explosive as a sneeze or just a simple misstep walking down stairs. The ability of individuals to return their trunk to another safe equilibrium condition following a perturbation depends on the active and passive mechanical responses of their trunk tissues. Different measures of trunk's response to perturbations (e.g., time to stop, maximum flexion, etc.) have been used as indicators of overall trunk stability (Dupeyron, Perrey, Micallef, & Pelissier, 2010; Grondin & Potvin, 2009; Hjortskov, Essendrop, Skotte, & Fallentin, 2005; Pedersen et al., 2007; Pedersen, Randers, Skotte, & Krustrup, 2009; Skotte et al., 2004). Since contributing aspects of trunk neuromuscular system (i.e., passive aspects, and reflexive and voluntary aspects) to the trunk stability could be affected differently by exposure to LBP risk factors or following a treatment or training protocol, recent research efforts have been focused to investigate the relative contribution of these different aspects of trunk behavior to trunk stability (Bazrgari et al., 2011; Cholewicki et al., 2005; B. D. Hendershot, Bazrgari, Nussbaum, & Madigan, 2012; Hodges, van den Hoorn, Dawson, & Cholewicki, 2009; Miller, Bazrgari, Nussbaum, & Madigan, 2013; Reeves, Cholewicki, & Silfies, 2006; Zazulak, Hewett, Reeves, Goldberg, & Cholewicki, 2007). As a result, many different types of perturbation systems have emerged that can be used to study lower back mechanical behavior. At the HMBL lab, we were interested in developing a system capable of a comprehensive study of lower back mechanics. In particular, the system was required to be able to measure the active behaviors of trunk tissues separately from the passive properties. Furthermore, the reflexive and voluntary aspects of the active behaviors were needed to also be separately measured. Therefore, the existing perturbation systems and their capabilities were needed to be reviewed to develop an optimal perturbation design that can satisfy these requirements

2.2 Review of Different Perturbation Systems and Testing Protocols

Following an intensive review of literature, seven different systems were identified from institutions across the globe, each with a different set of means for perturbing the trunk and obtaining kinematics and kinetic data. Providing availability of information, the following aspects of each system will also be discussed: safety, accuracy, independent and dependent variables, and research questions for which they were utilized.

2.2.1 System #1: This system was mainly used in research conducted within the Departments of Kinesiology at the University of Windsor and McMaster University (Brown, Haumann, & Potvin, 2003; Grondin & Potvin, 2009) where the investigators created the perturbation method shown in Figure 2.1. In this design, perturbations were applied by dropping a weight into a container held by the study participants. To perform perturbations, the participants would stand unrestricted on a force platform holding a lightweight plastic bin in front of their bodies with their elbows bent approximately 45°. A curtain separated the subject from an overhead cable-pulley system that hung a 5 kg (≈ 11 lbs) weight 2.5 cm (≈ 1 in) above the center of the plastic bin held by the participant. The curtain was used to remove any visual indication that the weight was going to drop thus preventing any anticipatory activation of the trunk muscles. It is unclear though if auditory clues were inhibited during perturbation tests. During experiments, the participants were usually instructed to relax until the weight was dropped into the bin then they were to return quickly to their original upright posture. Through this method of trunk perturbation, the investigators can control the magnitude of force applied to each participant, the time at which the perturbation begins, and whether or not the participant is aware of this time (i.e., independent variables). Dependent variables included ground reaction forces/moments, electromyography (EMG) of select trunk muscles, and trunk rotation in the sagittal plane. The ground reaction forces as well as the moment about the medial-lateral axis were obtained from a force plate system. Using an EMG system, the muscle activity of targeted muscles across the perturbation were measured, including the baseline activity beforehand. Lastly, a goniometer was used to measure the trunk angular displacements.

This perturbation system has also been used to conduct sudden unloading perturbations (Brown et al., 2003). To achieve this, rather than dropping a weight into the plastic bin held by the participant, the weight was removed either voluntarily by the participants or unexpectedly by the experimenters. For the voluntary method, the participant would release the weighted box onto a padded table as the experimenter counted down to one. Alternatively for the unexpected method, which could be performed at a known or random time, the experimenter pulled the weight up from the box via the pulley system as quickly as possible. The investigators would be able to use the same systems to measure their respective responses, except the participant would be inclined to “fall” backwards instead of forwards.

This perturbation system has been used to investigate the effect of back and abdominal muscle fatigue on trunk reflexive and voluntary mechanical responses (Grondin & Potvin, 2009). The major limitation of this method is that the actual applied perturbation to the trunk will depend on the mechanical impedance of the elbow and shoulder joints and would likely vary among different individuals. The risk associated with this setup seems to be minimal although it includes a potential risk for fall. Although participants were instructed not to move their feet during the perturbation, they were allowed to do so if it was needed to prevent them from falling. This method of trunk perturbation is beneficial because of its low cost, simplicity, and short set up time, however, it offers limited accuracy and experimental versatility for our research concern (section 2.1).

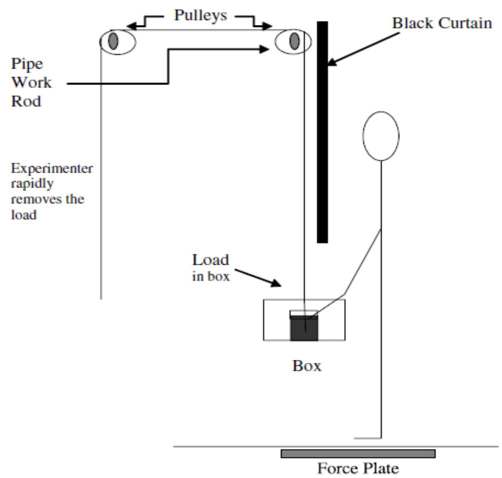

Figure 2.1 (Brown et al., 2003): The setup for perturbation tests at the Departments of Kinesiology at the University of Windsor and McMaster University. The sudden loading/unloading occurs when the load is dropped/removed into/from the participant's box.

2.2.2 System #2: The second design (See Figure 2.2) was primarily used by the National Institute of Occupational Health in Copenhagen, Denmark (Andersen, Essendrop, & Schibye, 2004; Essendrop, Andersen, & Schibye, 2002; Hjortskov et al., 2005; Pedersen et al., 2007; Pedersen et al., 2009; Skotte et al., 2004). Standing participants were strapped to a rigid structure at the waist and harnessed to a rod at the upper back. The rod was connected via wire and pulley to a 0.5 kg (1.1 lbs) weight to maintain wire tautness without inciting any noticeable activation in the participants' trunk muscles. The pulley and the weight were separated from the participant by a wall to prevent any visual clues of the impending perturbation. Also on the far side of the wall was a 5.9 kg (13 lbs) apparatus that was connected to two rigid structures via magnets. The apparatus also contained a gripping device and a solenoid to activate it. In order to generate the perturbation, a computer activated the solenoid causing the gripping device to clamp onto the cylindrical 0.5 kg weight, and then deactivated the magnets, releasing the apparatus from the fixed structure. This sudden vertical load was applied horizontally to the participant's trunk at the rod via the wire and pulley, thus causing the participant to bend forward about the waist strap (Skotte et al., 2004). The trunk muscles could be pre-activated beyond the level needed to hold the 0.5 kg wire weight by having the participants hold a weight in their hands (Andersen et al., 2004; Essendrop et al., 2002) or extending the trunk backwards a certain level of force before the perturbation (Hjortskov et al., 2005). Similar to the method used in the previous design, the participants were instructed to stand relaxed and upright until the perturbation occurred and then return to the upright posture as quickly as possible. The investigators were able to control the force applied to the participant by changing the mass of the apparatus. They were also able to control the timing of the sudden load and whether or not the participant was aware of the timing. During tests, trunk kinematic response was measured via a potentiometer attached to the pulley and the applied force to the participant via a force transducer located between the rigid bar of the harness and a plate fastened at the participant's back. Muscle responses were also measured using an EMG system. Other tests that were done using this system include standing maximum voluntary extensions (MVE) efforts and short latency stretch reflex. The short latency stretch reflex test required the participant to stand upright in the system and using an automatic reflex hammer the investigators produced quick taps on the erector spinae while measuring the muscle responses along with either the kinetics or kinematics data depending on the presence of the attachment to the loading apparatus (Hjortskov et al., 2005).

This design has been used to investigate the effects of training protocol on trunk mechanical behaviors. For instance, one experiment (Pedersen et al., 2009) placed 46 women into three groups: a soccer-training group, a running group, and an untrained group. Individuals in the training groups were trained for one hour twice a week for 16 weeks. The perturbation tests were performed before and after the training period comparing the different groups on stopping time and stopping distance following the sudden loading. One major limitation of this system is the oscillation of the applied force due to the elasticity in the cable and the bar in the harness (Skotte et al., 2004). This system, however, does seem to be safe as there was no risk for a fall, and if the weight proved too much for the participant to withstand, the gripping device would have released the weight and there was a platform to prevent the weight from going past a certain distance. Although experimenters only used this system in the forward bending direction, one could easily rotate the direction the participant was facing to perform perturbations in the lateral or the backwards bending directions. Despite the system's potential versatility, safety, and trunk isolation, the inaccuracies due to the lack of rigidity in the system make it fall short of our stated requirements described at the beginning of this chapter (section 2.1).

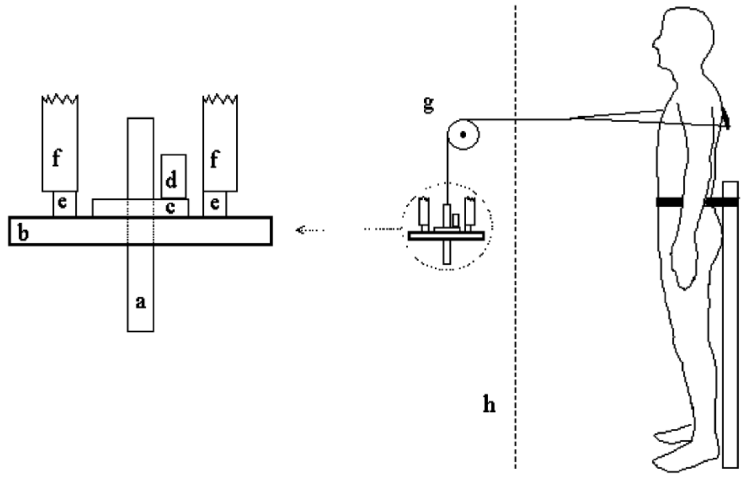

Figure 2.2 (Skotte et al., 2004): The perturbation system at the National Institute of Occupational Health in Copenhagen, Denmark. The loading device: (a) cylinder, (b) load, (c) gripping device, (d) activation solenoid for the gripping device, (e) holding magnets, (f) load-bearing construction, (g) reel with potentiometer, and (h) visual shield. The sudden perturbation occurs when the gripping device (c) grips the cylinder (a) and the holding magnets (e) are released.

2.2.3 System #3: This design (Dupeyron et al., 2010) has been developed and used by the Faculty of Sport Science and the Nimes Teaching Hospital in France (Figure 2.3). To begin, the participants were placed into a stationary fixture, restraining the lower extremities by pads at the knee/tibia, pads behind the thighs, and a seat belt around the hips. A harness was placed around the participant's chest at the T10 vertebra and attached to a strain-gauge type dynamometer, which was used to measure MVEs. For perturbations a large padded pendulum was set behind the participant at the T10 level and pulled back far enough to accumulate 50% of the total body mass in applied force when striking the participant. This should have eliminated all visual clues of the ensuing perturbations. (It is unclear, though, if there were any auditory clues to the movement of the pendulum.) The participants were instructed to return to neutral position immediately following the perturbation. During perturbations the investigators are able to control the mass of the pendulum, ultimately controlling the force applied to the participant. This applied force was measured using a strain gauge located inside the padded part of the pendulum. They were also able to control the time at which the perturbation would begin and whether or not the participant was aware of the start time. Dependent variables included EMG of select trunk muscles (i.e., external oblique and the erector spinae) and trunk motion.

This set up has been used to study the effects of trunk muscle fatigue on reflexive trunk mechanical behavior. Specifically, ten healthy men were tested before and after a back muscle fatiguing protocol. MVE efforts, reflexive latencies of external oblique and erector spinae muscle, normalized peak amplitudes of the erector spinae activity, and the level of co-activation between the external oblique and erector spinae were quantified and compared. It is unclear if there is a support at the front of the participant's pelvis or if the knee pad could provide enough support to restrict the pelvis from moving forward during perturbations without triggering any stabilizing muscle activity. It is also unclear what happens to the pendulum after it first strikes the participant. If it is completely free-falling then it will most likely hit the participants multiple times during each trial and would be pressing on their trunk as they try to return to neutral position. A major safety concern is striking the participants in the spine with 50% of their total body mass. An earlier modeling study of trunk response to sudden loading has reported spinal loads in excess of 5 kN at the L5-S1 level when an unexpected load of 100 N was applied to the trunk (Bazrgari, Shirazi-Adl, & Lariviere, 2009). Even if there were no immediate injuries to the ten young men in this study, the spectrum for eligible participants is very exclusive as overweight, elderly, and/or less physically fit people might not be able to withstand many of those high force impacts to the trunk. Although this perturbation system provides a structure that can successfully isolate the trunk muscles, it is limited by its safety concerns as well as its one directional perturbation and MVE tests.

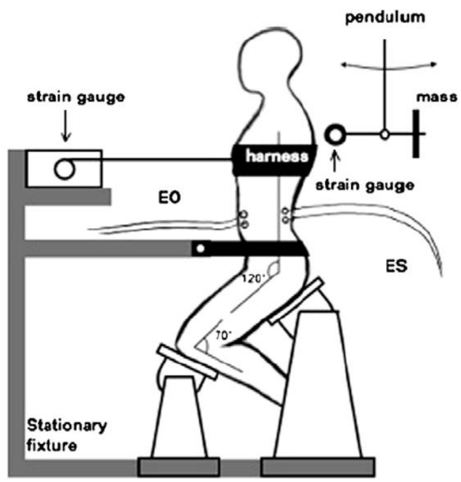

Figure 2.3 (Dupeyron et al., 2010): The perturbation system used at the Faculty of Sport Science and the Nimes Teaching Hospital in France. The mass at the end of the pendulum is lifted and swings into the back of the participant causing the sudden perturbation.

2.2.4 System #4: The fourth design (Miller, Slota, Agnew, & Madigan, 2010) has been used by researchers at the Virginia Tech and Pennsylvania State University (Figure 2.4). The participants stood with their legs against a rigid structure that is strapped to them around their pelvis. A harness containing a rigid rod was placed on their torso so that the rod was between the T6 & T8 level. A Kevlar cable attached the rod to a pendulum in front of the participants. The participants were instructed to hold a relaxed upright posture and close their eyes as the investigator released the weighted pendulum from a mechanical stop. The Kevlar cable became taut as the pendulum reached its final position at the vertical point causing one sudden load impact on the participants. With this system the investigators are able to control the weight of pendulum and the resultant applied force, the timing of the pendulum release, and the participants' knowledge of the release timing. The applied force was measured using a load cell located between the pendulum and the Kevlar cable and trunk kinematics was measured using a tri-axial Inertial Motion Sensor (Xsens, Culver City, CA). Muscle responses were also measured using an EMG system. As an example of studies that were performed using this system, one experiment compared the maximum flexion velocity and reflex latency with varying impulse magnitudes between men and women. This perturbation system appears to be both accurate and safe. The pendulum produced a single quick force impulse which removed the safety concerns of applying a constant load. Other tests that could be performed on this system include maximum voluntary extensions (MVEs) (assuming the mechanical stop at the pendulum's vertical position was sturdy enough to allow it) and perturbations in the backwards and lateral bending directions, similar to the tests performed at Yale University School of Medicine and Michigan State University (section 2.2.5).

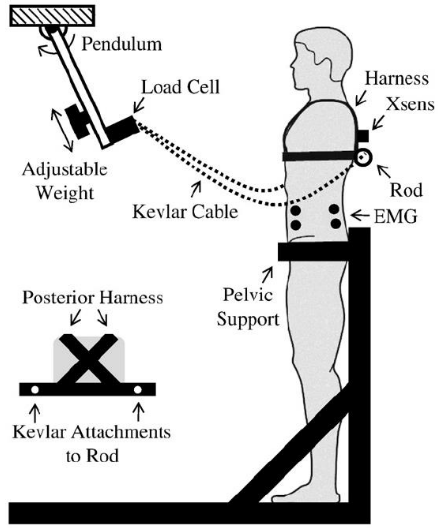

Figure 2.4 (Miller et al., 2010): The perturbation system used at the Virginia Tech and Pennsylvania State University. The sudden perturbation occurs when the weight at the end of the pendulum swings and pulls the participant via the Kevlar cable.

2.2.5 System #5: This design (Cholewicki, McGill, Shah, & Lee, 2010; Cholewicki, van Dieen, Lee, & Reeves, 2011; Hodges et al., 2009) primarily used by researchers at Yale University School of Medicine and Michigan State University (Figure 2.5). This design allows for perturbation in six different directions: bending forwards & backwards, left & right lateral bending, and twisting clockwise & counterclockwise. Regardless of the direction of perturbation, the participants were situated in the structure the same way. There were pads at the knee, behind the thighs, in front of and behind the hips to keep the participants rigid and isolate their trunk. The harness on the torso was placed on the spine between the T5 & T9 levels, the sternum, or the shoulders depending on the perturbation direction (Cholewicki et al., 2010). Weights were placed at the end of a cable that travelled across a pulley system and attached to one of the six locations on the harness. With the exception of configuration E (See Figure 2.5.E), in which weights were placed on multiple sides of the participant, the agonistic muscles must have been activated in order to counteract the weight and maintain an upright posture until the magnet release on the cable was activated dropping the weight. Once the weight was released the agonistic muscles were expected to reflexively relax, meanwhile the antagonistic muscles were expected to be reflexively activated to stop the motion and return the body to its original upright position. In the case of configuration E, the agonistic muscles were not required to be activated, assuming the weights on the front and the back were equal (Cholewicki et al., 2010). Once either weight was released the participant then needed to reflexively activate the appropriate muscles to stop motion, return to the original upright position, and maintain that position (Hodges et al., 2009). The activation/deactivation of these muscles was monitored by an EMG system and the trunk kinematic response was measured using a three-dimensional electromagnetic motion-measurement device (Cholewicki et al., 2010). The applied force was measured through a force transducer between the cable and the weight/rigid structure. This perturbation system has also been used for other tests. For example, in one study (Cholewicki et al., 2011) the cable was fixed to a rigid structure with an inline load cell in order to perform isometric MVCs in all six directions (See Figure 2.5.F). In any given direction the investigators were able to control the amount of weight applied to the participant, the timing of the perturbation, the participants' knowledge of the timing, and in configuration E the participants' knowledge of the perturbation direction. One experiment (Cholewicki et al., 2010) tested participants following a three-week use of lumbosacral orthoses. This system appears to be safe given that the variable weight can always be released and the body cannot fall out of the structure. Also, these force controlled perturbation methods seem to be accurate since the body is held rigid and the trunk muscles are well isolated. However, this system is not capable of separating the reflexive from the voluntary response, one of our research requirements (section 2.1).

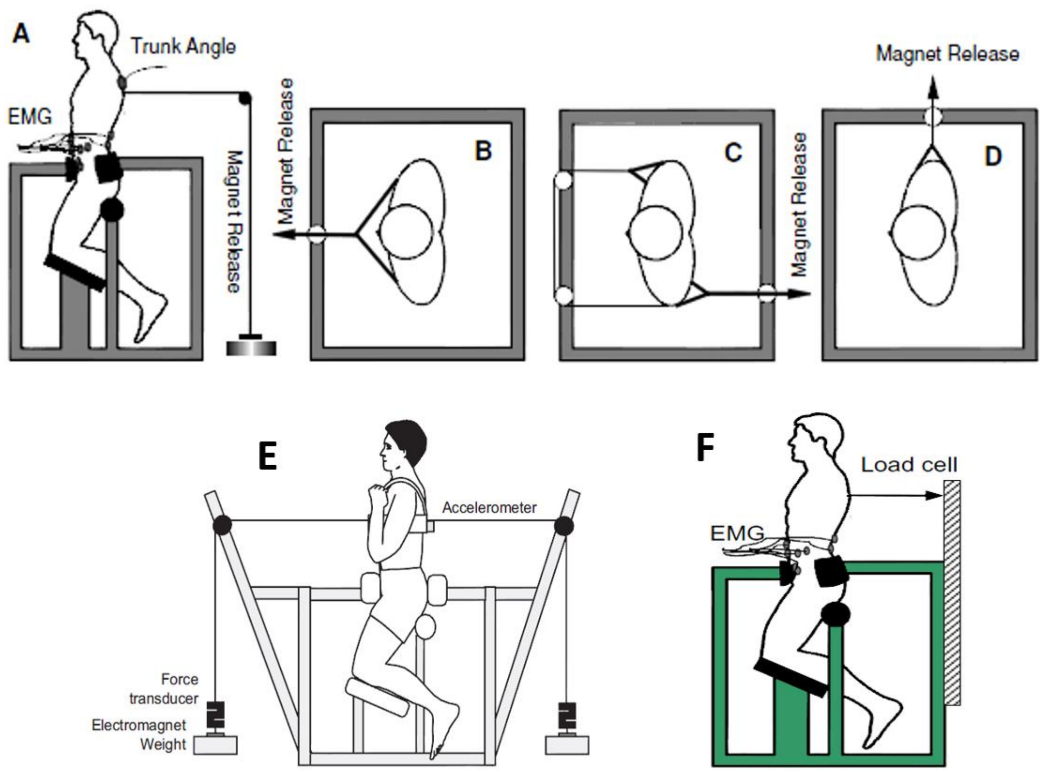

Figure 2.5 (Cholewicki et al., 2010; Cholewicki et al., 2011; Hodges et al., 2009): The perturbation system used at Yale University School of Medicine and Michigan State University. Sudden perturbations occurred when the magnet was released in a variety of setups: (A) forward bending, (B) backwards bending, (C) twisting, (D) lateral bending, and (E) forward or backwards bending without the pre-activation of agonistic muscles. (F) shows an isometric MVF.

2.2.6 System #6: This design (Bazrgari et al., 2011; B. Hendershot et al., 2011; B. D. Hendershot, Bazrgari, & Nussbaum, 2013; B. D. Hendershot et al., 2012; Miller et al., 2013; Toosizadeh et al., 2013) comes from the Department of Industrial and Systems

Engineering and the Department of Engineering Science and Mechanics both at Virginia Tech (Figure 2.6). Before perturbations began in this system, participants stood on a platform, strapped in at the waist. This platform was previously adjusted to align the participant's L5/S1 joint with the axle with which the platform can be rotated about via an electric linear actuator. A harness made from a chest guard and four metal channels were bolted together around the participants' torso (B. D. Hendershot et al., 2012). This harness was attached to an adjustable rigid rod at the front, left, or right side of the participant (Figure 2.6.B) that contained a load cell (B. D. Hendershot et al., 2013). This rod was attached on the other end to a servomotor that sat on top of an adjustable platform. Behind the participant and attached to a rigid structure on the frame were two laser sensors that measure the movement of the participants' trunk. Perturbations were performed as the motor alternately pushed and pulled the participants via the connecting rod. The investigators were able to control the frequency, velocity, acceleration, and distance of each perturbation as well as the total number of perturbations each participant experienced. The kinematic input, as controlled by the motor, was measured by the laser sensors and the motor's encoder. The variable force response was measured by the load cell in line with the connecting rod. This characterizes the force required by the motor to displace the participant. An EMG system was also used to measure the response of target muscles. One experiment (Miller et al., 2013) compared eight male athletes with recurrent acute LBP against nine male athletes with no LBP. In this experiment, the participants performed the perturbations while sitting on the raised foot platform instead of standing. The system has also been used to conduct three other tests:

1) isometric MVCs in the forward, backward, and lateral bending directions since the motor could fix the position of the rod; 2) stress relaxation tests by raising the participant's legs via the platform and actuator (Toosizadeh et al., 2013). The participant was instructed to maintain a relaxed upright posture for several minutes in the raised position while the load cell measured the resistance of the passive trunk tissues. 3) Creep deformation tests by removing the rod from the harness and placing weights around the participant's wrists (Bazrgari et al., 2011). The participant then bent over and relaxed for several minutes as the flexion angle slowly increased. Although not performed, this test could have been used in the second (section 2.2.2) and fourth (section 2.2.4) designs. The stress relaxation and creep deformation tests characterized the passive responses of the trunk tissues whereas the perturbation test characterized the active (voluntary and reflexive) responses of the trunk tissues.

This perturbation system seems to be safe as the participants were secured by the harness system which eliminate the risk of falling. There are also manual and electronic emergency stops that are either automatic or can be controlled by the participant and the experimenters to cut power to the motor at any time. This system also seems to be fairly accurate since the targeted trunk muscles are being isolated and the system between these trunk muscles and the motor is rigid. The displacement measurements at both ends of the system can account for any rigidity as well. This is the first example of a displacement controlled perturbation system, in which the displacement, velocity, acceleration, and timing of the perturbations are sent to the participant via the motor and the force is the variable output. This is also the first system that allows for consecutive perturbations in a small window of time. This system's versatility in different experiments that meet the criteria of separately measuring the active behaviors of the trunk tissues from the passive as well as the reflexive from the voluntary makes it the most ideal reviewed system.

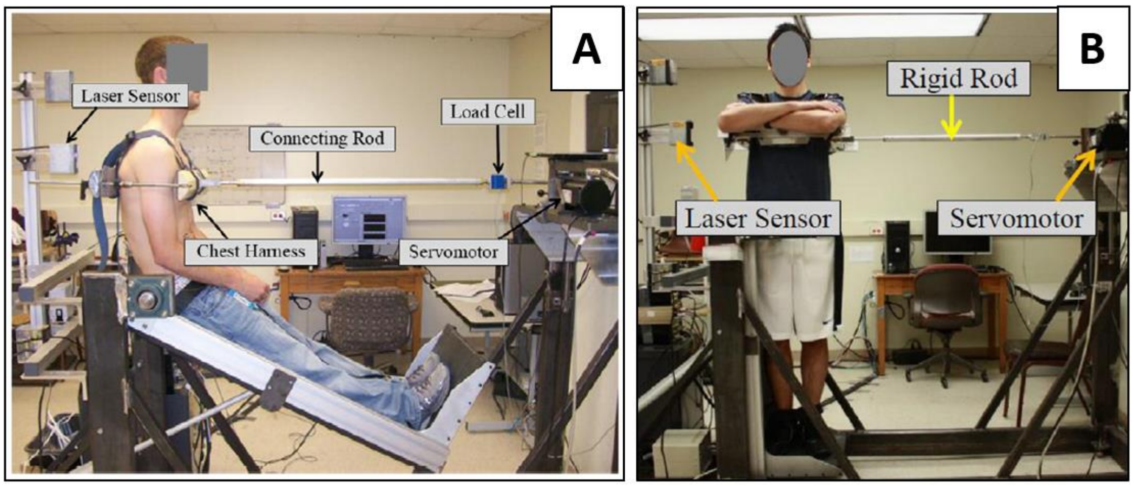

Figure 2.6 (B. Hendershot et al., 2011): The perturbation system used at Virginia Tech. A series of sudden perturbations are produced by the servomotor and applied to the participant's trunk via the connecting rod and harness.

2.2.7 System #7: The final design (Cort, Dickey, & Potvin, 2013) has been used by researchers at the Departments of Kinesiology at the University of Windsor, the University of Western Ontario, and McMaster University (Figure 2.7). The participants knelt on a circular robotic platform and were strapped in around the pelvis, thighs, and calves. The participants wore shoulder pads with added mass to increase the potential energy above the L4-L5 vertebrae, enhancing the effect of each perturbation. To produce lateral bend perturbations the robotic platform would quickly move linearly left or right 4 cm (left lateral bend via right platform displacements and vice versa). Other perturbation directions could be produced by the robotic platform by moving similarly in any of its six degrees-of-freedom. The investigators could control the direction, the participants' timing knowledge, and the participants' direction knowledge of the impending perturbations. An active marker system was used to collect kinematic data of the participants' trunk and a tri-axial accelerometer on the robotic platform to measure the accelerations and timing of the perturbations. An EMG system was also used to measure the response of trunk muscles. One experiment tested seven males in order to measure the trunk muscle forces and how they contribute to L4-L5 joint rotational stiffness. This system seems to be accurate and safe. Using the knelt position and the strap system, there is minimal risk for falls and the system should be usable by participants of all shapes and sizes.

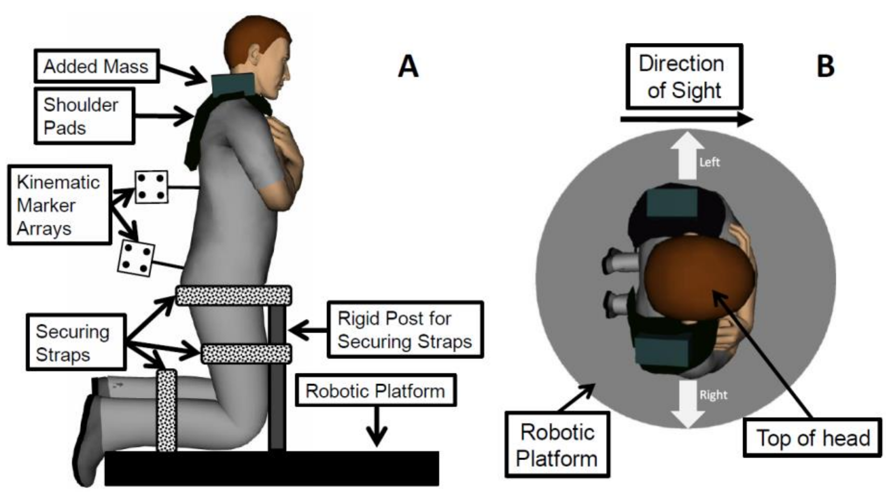

Figure 2.7 (Cort et al., 2013): The perturbation platform used at the Departments of Kinesiology at the University of Windsor, the University of Western Ontario, and McMaster University. Sudden perturbations occur when the robotic platform moves in the opposite direction of the intended perturbation causing forward/ backward bending, lateral bending, or twisting.

2.2.8 Summary and conclusion: We reviewed seven different classes of perturbation systems used to study lower back mechanical behavior. The sixth reviewed perturbation system from Virginia Tech was the only system which meets our research requirements. This system has successfully been used in the past to measure either the active or the passive trunk tissue mechanical behaviors. This design also provides the capability of separating the reflex response from the voluntary response. Moreover, this design offers the capability of performing many consecutive perturbations in a short amount of time which improves the overall data collection process.

2.3 HMBL Design

The perturbation system at the University of Kentucky (See Figure 2.8) was constructed primarily to the design at Virginia Tech, but the designs of the other systems were used to improve upon this system. The components of the perturbation system can be categorized into six groups: the frame, the leg platform, the motor platform, the harness, the connecting elements, and the electrical components.

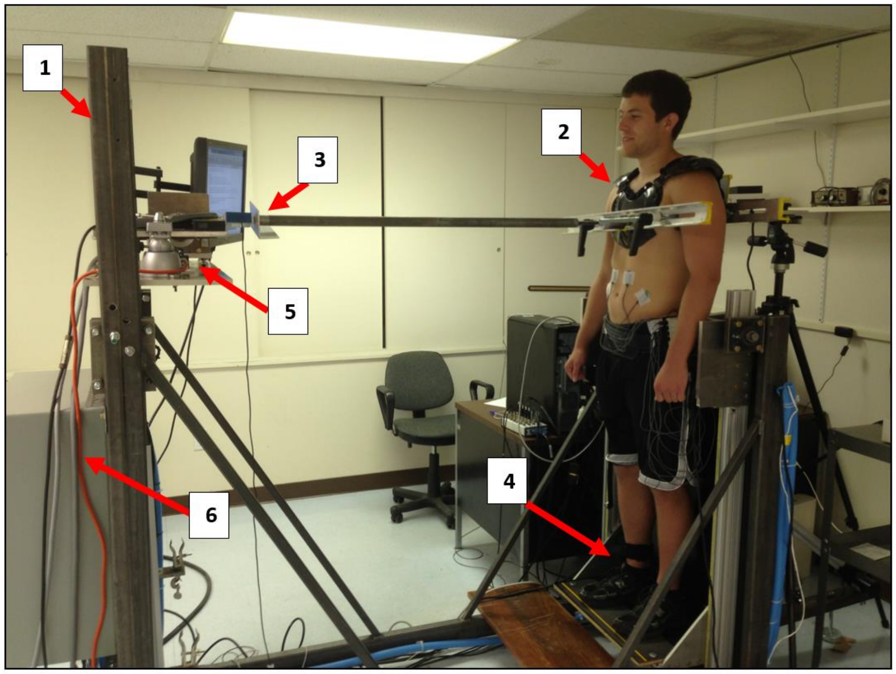

Figure 2.8: Perturbation system in the Human Musculoskeletal Biomechanics Lab at the University of Kentucky. Six groups of components: 1) the frame, 2) the harness, 3) the connecting elements, 4) the leg platform, 5) the motor platform, 6) the electrical components.

2.3.1 Frame: The frame (see Figure 2.9) consists primarily of steel rectangular tubes (3” x 1.5” x 3/16”). The base (length: 6 ft; width: 2.5 ft) is held up by four vibration-damping leveling mounts bolted each to a steel plate (4”x 3.5”x 3/8”) which were welded to the four inside corners of the base. The anterior side of the frame (height: 6 ft; width: 2.5 ft) supports the motor platform and the feedback monitor, while its posterior side (height: 4 ft; width: 2.5 ft) supports the leg platform via two flange bearings (The Big Bearing Store, Memphis, TN) bolted to the top of the posterior side of the frame. Anterior and posterior sides of the frames were welded to the base both directly and indirectly via steel tubes (1” x 1” 1/8”) for added support and rigidity. Although solid steel beams would've been better for the frame in order to reduce vibrational effects of the motor, the steel tubes were chosen because they were more cost efficient, easier to transport due to lighter weight, and still provide adequate resistance to vibrations.

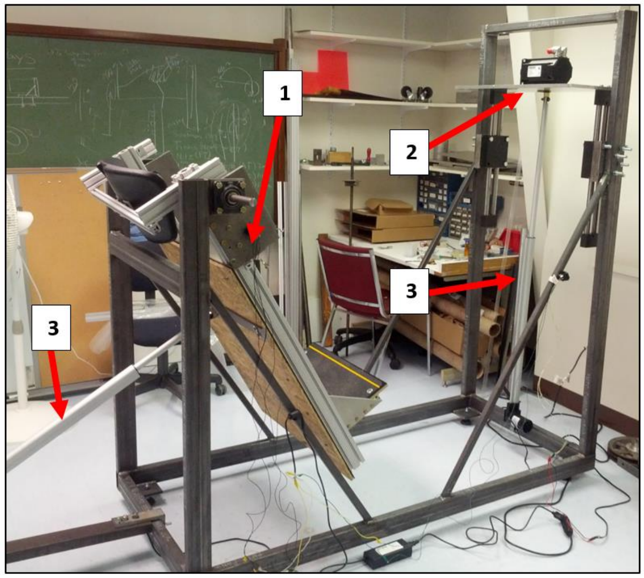

Figure 2.9: The frame provides a sturdy base to inhibit vibrations during perturbations. The anterior and posterior sides of the frame accommodate the motor platform (2) and the leg platform (1), respectively. Adjustability of the two platforms is provided by two separate linear electrical actuators (3).

2.3.2 Leg Platform: The leg platform was constructed using aluminum profiles, t-bolts, and t-nuts (See Figure 2.10). The two aluminum profiles (49" x 3" x 1.5") were used to provide adjustability for height, enabling accommodation of individuals with different leg heights. The leg platform connects to the frame via two aluminum plates (7" x 10" x ½") and two stainless steel rods (diameter: 7/8 in; length: 7.5 in) at the two flange bearings. The backside of the frame of aluminum profiles was bolted to an oriented strand board (OSB) (36"x 29" x 7/16") behind an equally sized plastic board used for aesthetics and comfort. A seat from a swivel chair was bolted to the frame (length: 12 in; width: 29

in) above the boards. The seat served as a restraint to minimize hip rotation as the leg platform was raised. The distance of the seat from the frame center of rotation is also adjustable. A standard car seat belt (Seatbelts Plus, Oceanside, CA) was t-bolted to the frame, which, along with the seat, restrains translational movement of the subject's pelvis as well. In front of the boards runs a foot rest (length: 12 in; width: 22 in), which was t-bolted to the inside of the aluminum frame near the bottom. The foot rest is a frame (length: 10 in; width: 22 in) made of aluminum profiles (1" x 1") bolted together by four aluminum plates (3" x 4.5" x ¼"), one at each corner. Bolted above the foot rest is a pair of wood (12" x 22" x ½") and plastic (12" x 22" x 1/8") boards with a tread strip adhesive to prevent slipping. An aluminum right triangle (10" x 10" x ½") was t-bolted to each side of the foot rest and t-bolted to the inside of the aluminum frame. In order to account for the different height of participants, the location of foot rest along the vertical profiles needs to be adjustable. Therefore instead of using standard t-bolts to connect the triangles to the profiles, adjustable handled bolts (McMaster-Carr, Aurora, OH) were used to allow for quick adjustability. Additionally, two measuring tapes were cut and taped to the outside of the vertical profiles to easily mark the 44 inches from the axles to the foot platform. On the back side of the leg platform, also bolted to the vertical profiles of the aluminum frame is a steel bar (29" x 2" x ¼") which connects the leg platform assembly to the actuator. The 400-lb capacity linear actuator (Firgelli Automations, Surrey, BC) was pinned to half of a steel I-beam (4" x 2.75" x 1.5") which is bolted into the leg platform boards through the steel bar and controlled by a "momentary" rocker switch (Firgelli Automations, Surrey, BC). The other end of the actuator (i.e., the motor side) was similarly pinned to an identical I-beam half that is bolted to a steel rectangular tube (3” x 1.5” x 3/16”; length: 33”). This tube was bolted to the frame using a steel L-bracket (length: 7 in; height: 4 in; width: 2 in; thickness: ½ in). A leveling mount (McMaster-Carr, Aurora, OH) was also bolted to the tube to prevent movement while the actuator is in motion. With this setup, the leg platform is able to be rotated around its connecting points at the perturbation system frame up to ≈ 90°. Apart from the motion measurement instruments, rotation of the leg platform can be monitored by a circular protractor adhered to the outside of the axle that rotates with the platform and a stationary nail adhered to the frame.

Figure 2.10: The foot rest can quickly be adjusted along the leg platform to accommodate a wide range of participant heights.

2.3.3 Motor Platform: An AC synchronous brushless servomotor (Kollmorgen, Radford, VA) was selected for the generation of displacement controlled perturbations. The servomotor has a peak torque of 26.5 Nm (235 lb-in) and max speed of 1740 RPM. It sits on and is bolted to a platform made of an aluminum plate (24”x 18”x 0.5”). This motor platform (see Figure 2.13) likewise sits on and is bolted to two vertical slider guides on each side and one linear actuator (Firgelli Automations, Surrey, BC) in the center. The vertical slider guides are each bolted to the anterior sides of the frame via large aluminum blocks which are bolted to the inside of the upright frame and allow the motor platform to move vertically but not laterally. The slider guides allow for 20 inches of vertical adjustment. This adjustability plus the adjustability of the leg platform that the participant stands on provide the complete range of heights for viable participants. The base of the linear actuator is bolted to a beam in the base of the frame and is controlled by a separate “momentary” rocker switch (Firgelli Automations, Surrey, BC). Between the motor platform and the motor is a pair of one-inch aluminum spacers. They aid in the connection of the motor to the platform as well as provide the required height for devising a limit switch. The limit switch (McMaster-Carr, Aurora, OH) is used to kill all power to the motor if the motor ever rotates more than 70° in either direction past the center, which means the subject will not experience greater than 1.4 inches of horizontal displacement. It requires the experimenter to manually reset it before the motor can be powered again.

2.3.4 Harness: The harness (see Figure 2.11) was made from a R3 roost deflector (Fox Head, Irvine, CA) that is used in motocross to protect the rider from flying debris. This type of deflector is ideal for use as a harness because it is sturdy and can be tight on the user without sacrificing movement or comfort. The harness distributes the impact from the servomotor over the thorax equally. Two U-channels (length: 27 in) were bolted to the harness horizontally; one in the front and one in the back at the sternum level. Each

U-channel had two slots (length: 8 in; width: 0.5 in) to pass two ½-13 threaded rods (length: 16 in) between the U-channels on both sides of the user. Each rod could be tightened independently with quick-clamp adjustable handles. Also bolted to the rear of the harness was a smooth plastic plate. This was used as a flat surface for the laser displacement sensor (section 2.3.6) to prevent errors in the displacement reading due to the contour of the harness or the participants back. The harness system is attached to the connecting elements using a quick release system. The quick release system comprises of an internal solid aluminum shaft (diameter: 0.5 in; length: 2 in) bolted to the harness and inserted into a hole (diameter: 0.5 in; length: 1 in) in an aluminum cylinder (diameter: 1.5 in; length: 3 in) coupled via a quick release pin through designated holes (diameter: 0.25 in) in both cylinders. Such design is important for a rigid connection of the harness with connecting elements while providing a quick release option in case of emergency. We fabricated two harness systems to accommodate a wider range of statures.

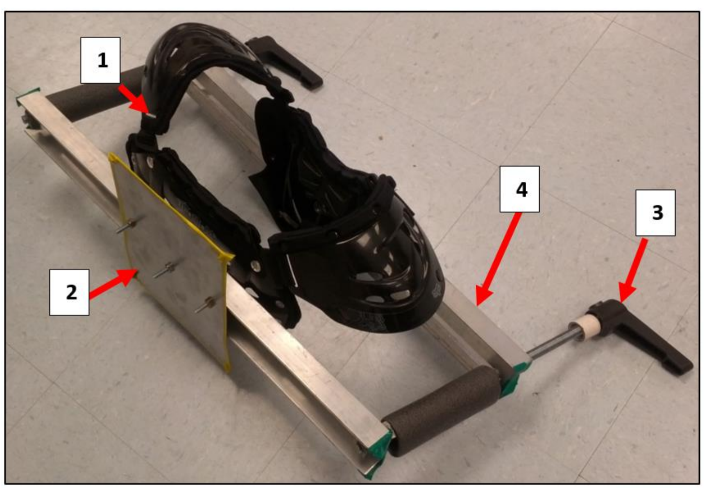

Figure 2.11: The harness consists of a (1) roost deflector, a (2) plate for the laser sensor, (3) quick-clamp adjustable handles, and (4) two rigid U-channels.

2.3.5 Connecting Elements: This assembly (see Figure 2.12) begins with the cylinder components of the harness' quick release system. The main component of the connecting elements is a steel pipe (length: 38 in; diameter: 1.25 in) which on one end is connected to the harness via the quick release system and on the other end is connected to the servomotor via a load cell and a crank mechanism. Starting with the end of the quick release system, there is a 12mm tapped-hole which connects the quick release system to the steel pipe via a 12mm threaded rod (length: 3 ft). The connection of the threaded rod to the steel pipe was facilitated using a 12mm hexnut welded to each end of the pipe. Such design allows for adjustment of the connecting element length (~±1.5 ft of adjustability in each direction) to accommodate different body statures. Length adjustment is achieved by rotating the threaded rod in or out of the steel pipe. The other end of the pipe is rigidly connected to another 12mm threaded rod (length: 4.5 in) which is also rigidly connected to a load cell (Interface SM2000, Scottsdale, AZ). In front of the load cell's lock washer and nut there is an L bracket (length: 9 in; height: 3 in), which is used as the target for a laser displacement sensor to measure the load cell's displacement during perturbations. An L bracket is used here instead of a simple plate to reduce the amount of wobble and vibration that occurs during perturbations.

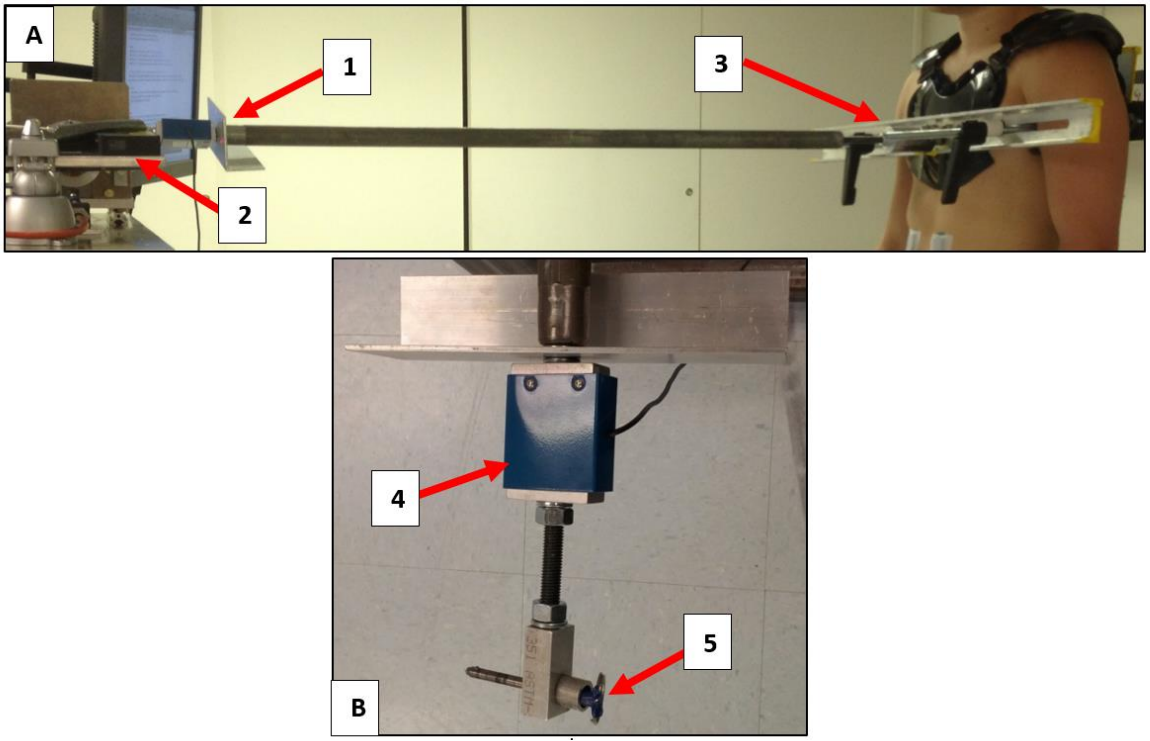

Figure 2.12: The (1) L-bracket is used by the (2) laser sensor to measure the kinematics of the (4) load cell. The (3) quick-release system connects to the harness and the (5) quick-release pin connects to the cylinder component of the crank mechanism.

The other end of the load cell is connected to the servomotor via a crank mechanism. The crank mechanism includes a 12 mm threaded rod, which at one end is connected to the load cell, and at the other end is connected to a coupling block which connects the threaded rod to a cylinder via a quick release pin. Finally, the cylinder part of the crank is coupled with the servomotor using a center hole and a keyway. The cylinder component of the crank mechanism (see Figure 2.13.B) contains many different features in order to interact with many different parts of the system. The arc at the bottom will strike the limit switch if the motor rotates more than 70°. The quarter-inch hole at the top allows the connecting rod to attach via a quick-release pin while still allowing for rotation and easy reattachment. The center hole & keyway allows for coupling of the crank mechanism with the motor. Finally a second quarter-inch hole is located on the outer rim of the cylinder to allow for a quick-release pin to attach and serve as an additional brake for the motor.

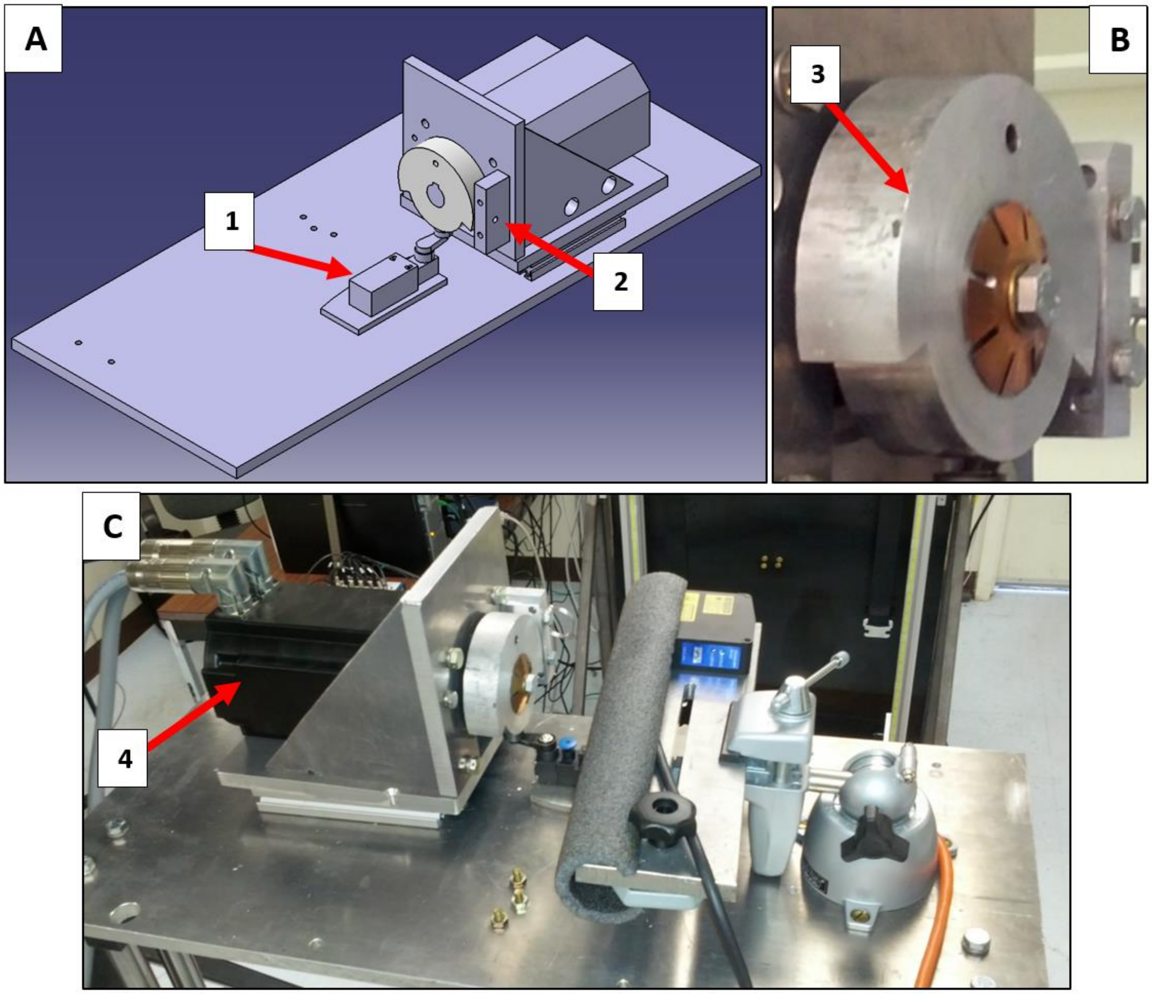

Figure 2.13: The (1) limit switch will cut power from the (4) motor if it is triggered by the (3) cylinder component of the crank shaft. The (2) quick-release pin brake prevents the motor and the participant from moving.

2.3.6 Instruments and Electrical Components: The perturbation system was equipped with instruments to measure the kinematic and kinetic responses of the participants during the experiments. Kinetic response is measured by the in-line load cell on the connecting element assembly. The load cell measures both tension and compression with a capacity of 2000 N (~450 lbf). Kinematic response is measured using two laser displacement sensors (Optex-FA, West Des Moines). One is attached to the motor platform and is used to measure the location of the connecting elements at the load cell. The other one is attached to a camera tripod (Manfrotto, Italy) and is used to measure trunk displacement by targeting the attached plate on the back of the harness. These sensors have a measuring range of 85 ± 20 mm and a resolution of 1 µm. Each laser displacement sensor contains two tapped-holes and was bolted to an aluminum plate (12”x 3.5”x 0.5”). These plates were then used to facilitate connection with 1) the motor platform via a clamp (PanaVise, Reno, NV) and 2) the camera tripod via the tripod’s camera attachment system. A computer monitor was bolted to an articulating arm TV wall mount which was bolted on the upper right edge of the anterior side of the frame. Since the bracket swivels, it can always be in sight of the participant during experimental trials. The monitor is used to provide feedback to the participants.

In order to maintain a maximum amount of safety for the participants in the perturbation system, six emergency stops were implemented into the system. The first, as previously discussed, is the limit switch on top of the motor platform. This switch will cut all power to the motor if it rotates 70° past the normal. The second is a pendant switch (McMaster-Carr, Aurora, OH) that is controlled by the participant. Similar to the limit switch, if the button on the pendant switch is ever released, all power to the motor will be cut. This is proven to be a safer option than having the participant needing to push the button to cut the power, in cases where the participant faints or loses consciousness. There is a mushroom emergency stop button (Consolidated Electrical Distributors, Georgetown, KY) on the right edge of the posterior side of the frame near the axle of the leg platform. Although this button is within the reach of the participant, it is the primary way for the experimenter to cut power to the motor. On the right edge of the anterior side of the frame, right below the monitor's mount, there is an on/off switch that controls the power to the entire system. There is also a large switch located on the outside of a lock-out electrical box that controls power to the entire system. The lock-out electrical box contains the drivers for the motor and the laser sensors as well as the systems power control. Finally, there is an emergency stop control in the computer program that controls the motor. This would be the primary emergency stop for the experimenter controlling the motor.

The motor is controlled by the Kollmorgen Drive GUI (Graphical User Interface). This program manages the magnitude, direction, velocity, and timing of each perturbation. Matlab (Mathworks, Natick, MA) was used to control the remaining functions of the perturbation system. This includes data collection from the load cell and the two laser sensors and displaying the force response for the participant on the monitor.

2.3.7 Summary: The constructed system for the HMBL is capable of performing perturbations in the forward and backward bending directions at a controlled displacement, velocity, and acceleration at variable intervals. The perturbations are accurate due to the rigidity of the system, the isolation of the target trunk muscles, and the dual laser sensors measuring the input and output displacements. Although the perturbation system has been tested, generating comparable data with those from the system at Virginia Tech, the validity of the measurements from the system proved difficult to verify. This is due to the lack of known values of lower back mechanical properties to justify the results. The perturbations are also safe due to the numerous emergency stops as well as the sturdy straps to keep the participant in place. This system is also capable of performing accurate isometric MVE and MVF efforts as well as the stress relaxation and creep deformation tests due to the rigidity of the motor, the connecting rod, the harness and the leg platform. Performing any of these activities requires at least two operators.

CHAPTER 3: THE EFFECTS OF PHYSICAL ACTIVITY ON TRUNK MECHANICAL AND NEUROMUSCULAR BEHAVIORS

3.1. Introduction

Back pain is the second most common neurological ailment in the United States (US), being surpassed only by headaches (NINDS, 2004). Approximately 70-80% of US adults experience back pain at least once in their lifetime (Rubin, 2007). As one of the most common symptoms prompting physician visits, back pain can also be very costly (Deyo & Phillips, 1996). It has been estimated that individuals with back pain to spend 60% more on health care than individuals without back pain (Luo et al., 2004). Despite the high prevalence, only 10% of back pain cases have known causes and consist primarily of injuries and disease (i.e., spinal fractures, degenerative disc disease, cancer) that can be identified from a radiograph and treated as necessary. The remaining 90% of back pain cases are considered to be nonspecific since their causes are unknown (Manek & MacGregor, 2005). While the etiology of most back pain cases remains unclear, epidemiological studies have identified several occupational (physical and psychosocial) and non-occupational (i.e., personal) factors that are associated with a high incidence of low back pain (LBP) (Manek & MacGregor, 2005). Such knowledge of LBP risk factors can be used to understand its etiology via unraveling the underlying mechanism(s) which links exposure to risk factors with occurrence and recurrence of LBP.

Abnormal mechanics of the lower back, specifically stress and strain distributions that instantaneously or cumulatively exceed the injury thresholds of lower back tissues, have been suggested to be related to LBP occurrence (Cholewicki et al., 2005). Stress and strain distributions among the lower back tissues depend on the active and passive mechanical response of these tissues to the physical demands of the activity. Acute lab-based exposure to a single physical risk factor for LBP has been shown to cause alterations in both active and passive aspects of lower back tissues behavior that require a longer recovery time than exposure time (Bazrgari et al., 2011). Further, changes in aspects of lower back behavior following acute lab-based exposure to LBP risk factors appear to adversely affect the lower back mechanical environment. However, it is not clear whether a longer time of exposure to physical risk factors over the course of a work day could cause cumulative changes in aspects of lower back mechanics.

The purpose of this study was to address the above question by quantifying the effects of level of physical activity on diurnal changes and overnight recoveries in aspects of lower back mechanics. Using a newly developed set of computational and experimental tools, diurnal changes and overnight recoveries were obtained in two groups of individuals with respectively high and low levels of daily physical activities. It was hypothesized that individuals with a higher level of physical activity to experience greater alteration in aspects of trunk mechanical behavior that do not fully recover following the overnight rest period.

3.2. Methods

3.2.1. Participants

To determine the effects of level of physical activity on diurnal changes in lower back mechanics and overnight recovery, measures of lower back mechanics were collected from two groups of male participants: 1) four members of the University of Kentucky Club Ultimate Frisbee team (age: 21.5 (2.1) years, height: 185.4 (3.4) cm, and weight: 90.3 (9.0) kg) representing a group with high level of physical activity (HPA), and 2) six students and faculty (age: 26.8 (4.4) years, height: 178.1 (4.2) cm, and weight: 77.7 (9.3) kg) representing a group with low level of physical activity (LPA). During the course of this study, the member of the HPA group were participating in an ultimate Frisbee tournament that exposed them to at least six hours of intensive physical activity in one day that involved many repetitions of trunk motions (see Appendix). The LPA members, on the other hand, were only involved in their daily sedentary work routine and were instructed not to exercise or physically exert themselves between testing sessions. The exclusion criteria included a recent history of LBP, any spinal surgery in the lumbar region, and any current musculoskeletal disorder. Prior to screening, all participants completed a consenting procedure approved by the University of Kentucky's Institutional Review Board.

3.2.2. Experimental Procedures

Each participant completed three data collection sessions in two consecutive days. The first and second sessions were respectively completed in the morning (before exposure: BE) and the evening (after exposure: AE) of the first day and the third session was completed in the morning of the second day (after recovery: AR). In average (SD) participants had 7.6 (0.9) hours of sleep between the second (i.e., AE) and the third (i.e., AR) sessions. To minimize any residual effects on trunk mechanics due to physical activity from days prior to the first day of data collection, members of HPA group were tested during the first two days of the ultimate tournament. During each session, the participants stood on an adjustable platform in a rigid metal frame and were strapped in at the pelvis (Figure 3.1). A harness was placed on the participants' thorax to facilitate connection of participants, via a rigid bar at ~ the T8 level, to a mounted servomotor (Kollmorgen, Radford, VA) on the frame. The foot platform and the servomotor platform were adjusted for each participant such that the axle of the leg platform was at the participant's L5-S1 level, the trunk was upright, and the connecting rod was horizontal.

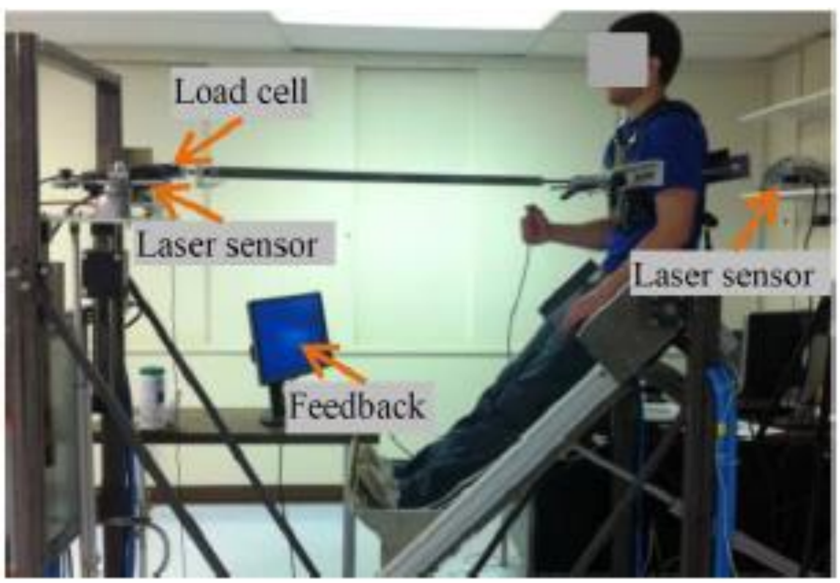

Figure 3.1: The leg platform is rotated to perform the stress relaxation test; the tension in the connecting rod is measured via the load cell.

While inside the experimental setup, each participant completed three sets of tests: 1) maximum voluntary exertions, 2) trunk perturbations, and 3) stress relaxation. For the maximum voluntary exertions the servomotor was locked and the participants were instructed to pull or push increasingly hard until their maximum and hold for two seconds. Separated by minute-long breaks, each exertion was repeated once for a total of four maximum voluntary exertion tests. During these tests, tension in the connecting rod was measured (sampling rate: 1000Hz) using a load cell (Interface SM2000, Scottsdale, AZ). Examples of this are shown in Figure 3.2.

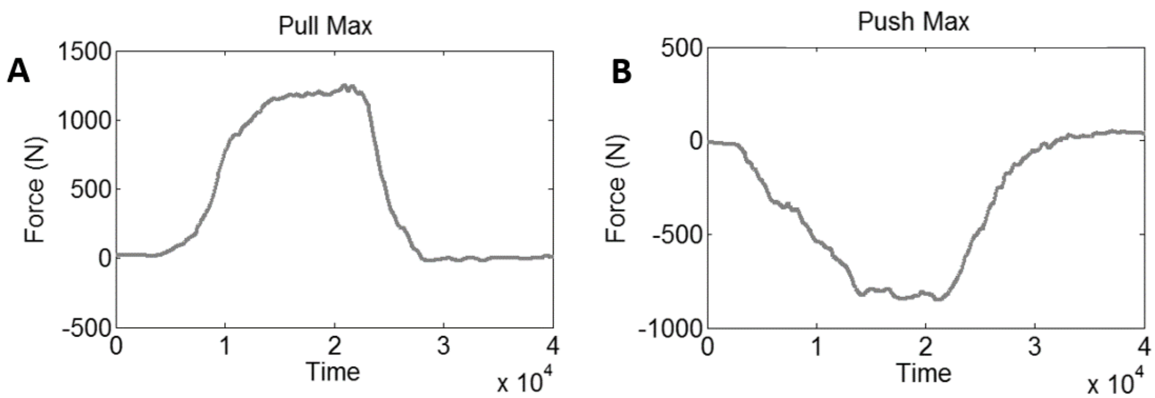

Figure 3.2: Sample recorded force by the in-line load cell during maximum voluntary contraction tests. Each (A) MVE / (B) MVF test was performed twice.

For the trunk perturbation tests, a series of pseudo-randomly time-spaced clockwise and counterclockwise angular displacements (± 8 degrees) were generated by the servomotor. This alternating angular displacement sequence was translated to horizontal back and forth motion (± 10 mm) using a crank mechanism and were applied to the participants' trunk via the harness-rod assembly. The tests were run at 34 full perturbations per minute for 30 seconds. Alterations in perturbation rate would be associated with changes in the model prediction of the lower back mechanical properties due to the viscoelastic properties of the lower back. However, the focus of the study is on relative changes rather than absolute values, therefore the results should not be affected by such rate dependent mechanical properties. During each perturbation test and using real-time force feedback, participants were instructed to hold 10% or 30% of their maximum pulling exertions (i.e. the average of the peak exertion forces). Trunk resistance to displacement perturbations was measured using the in-line load cell while two high-accuracy laser displacement sensors (Optex-FA, West Des Moines, IA; resolution: 1 µm; sampling rate: 1000 Hz), one targeting the participants' back at the T8 level and another targeting the load cell, were used to collect kinematic data. Examples of measured kinematics and kinetics data during the perturbation tests are shown in Figure 3.3.

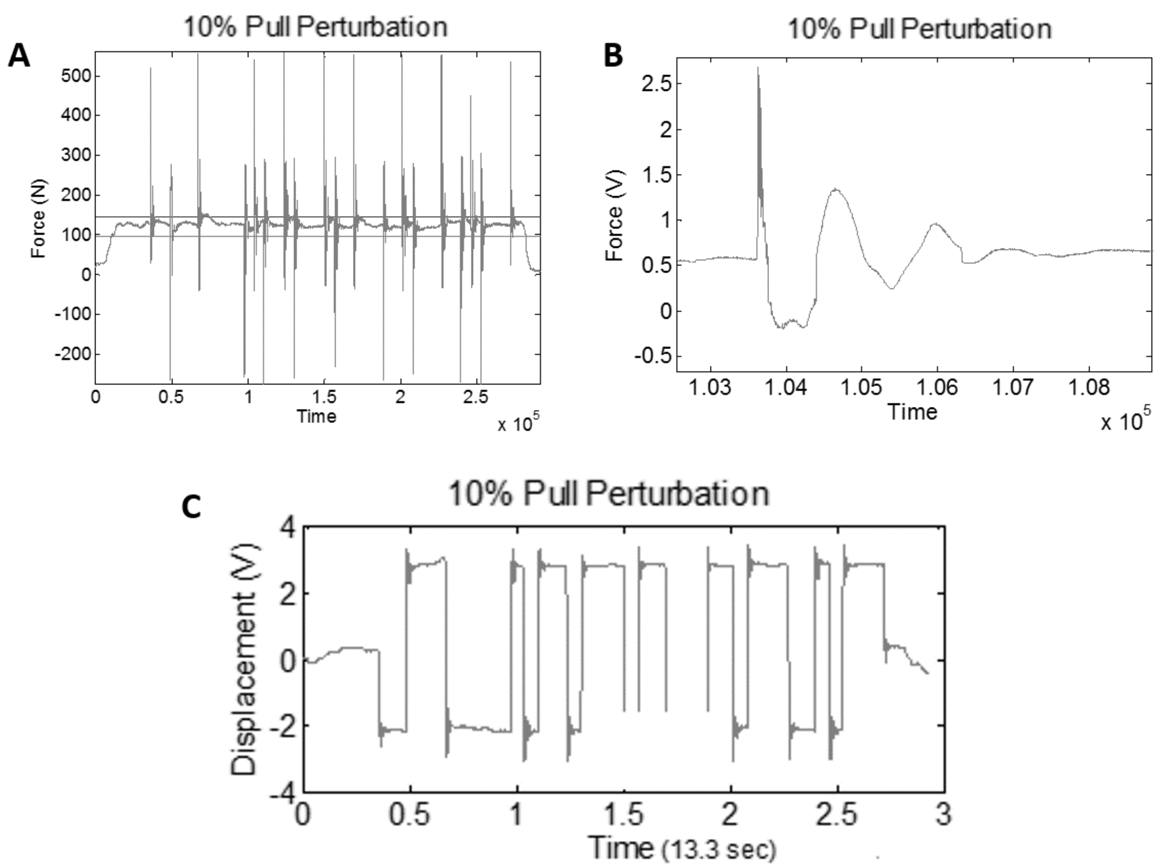

Figure 3.3: (A) During the perturbation tests, the participants were tasked with maintaining a pulling force between the two horizontal lines. (B) Following each perturbation was the lower back's reflex and voluntary responses. (C) Although the distance of each perturbation was the same, the interval between perturbations was pseudorandom.

Finally for the stress relaxation tests, while the servomotor was locked, the leg platform was rotated around its axle (i.e., aligned with the participants' L5-S1 joint) to generate ~ 40 degrees of flexion in the lower back. Since the trunk was in an upright posture and the demand in the active tissues was minimal, the recorded force by the load cell mainly represents the passive viscoelastic tissue resistance to the applied forward deformation (or flexion). The participants remained in this position for 4 minutes. An example of recorded force during stress relaxation tests is shown in Figure 3.4.

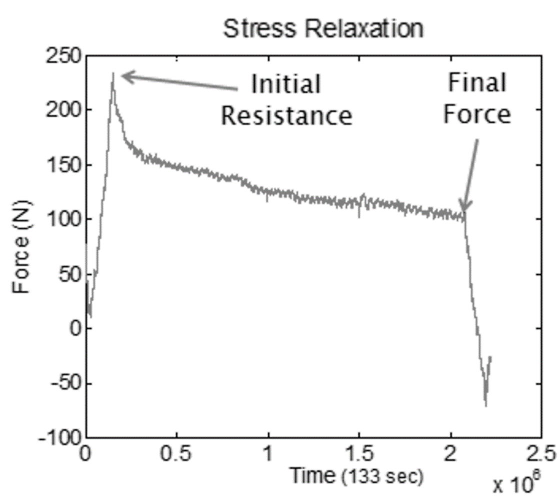

Figure 3.4: The initial peak resistance occurs once the lower back reaches 40 degrees of flexion. For four minutes the resistive force steadily decreases until the participant is lowered back to an upright posture.

3.2.3 Data Analyses

Maximum recorded forces during the two repetitions of pulling and pushing exertions were averaged and reported as measures of maximum voluntary trunk extension (MVE) and flexion (MVF) efforts. Intrinsic mechanical properties of the trunk were calculated similar to our earlier works (Bazrgari, Nussbaum, & Madigan, 2012). Briefly, the measured kinematics and kinetics data obtained during the trunk perturbation tests were used to characterize parameters of a two-degree of freedom mass-spring-damper system representing the trunk and connecting elements between the trunk and the servomotor (as shown in Figure 3.5). Similar to the assumptions made in our earlier works, the system identification procedure was constrained to zero damping. Hence, any alteration in trunk mechanical impedance was quantified by changes in trunk intrinsic stiffness and apparent mass. Finally, trunk passive viscoelastic resistance to 40 degree flexion was quantified by the magnitude of recorded forces at the onset (initial force: IF) and the end (final force: FF) of the four-minute exposure period of the stress-relaxation test. A repeated measure ANOVA was used to evaluate the effects of exposure level/group (i.e., independent factor with two levels: high/HPA and low/LPA) and time (i.e., repeated factor with three levels: BE, AE, and AR) on our measures of trunk mechanics. Considering Bonferroni adjustment on the significance level (i.e., afw = a/3 = 0.017), significant ANOVA results (i.e., $p < 0.05$) were followed by paired t-tests to determine the significance of diurnal changes as well as overnight recoveries. The effects of level of effort (i.e., 10% and 30% MVEs) on the measures of trunk impedance (i.e., intrinsic stiffness and apparent mass) were also investigated using analyses of covariates.

2-DOF model:

$$F(u,t) = \begin{Bmatrix} F_1 \\ 0 \end{Bmatrix} = \begin{bmatrix} 1.5 & 0 \\ 0 & m_2 \end{bmatrix} \begin{Bmatrix} \ddot{u}_1 \\ \ddot{u}_2 \end{Bmatrix} + \begin{bmatrix} c_1 & -c_1 \\ -c_1 & c_1 + c_2 \end{bmatrix} \begin{Bmatrix} \dot{u}_1 \\ \dot{u}_2 \end{Bmatrix} + \begin{bmatrix} k_1 & -k_1 \\ -k_1 & k_1 + k_2 \end{bmatrix} \begin{Bmatrix} u_1 \\ u_2 \end{Bmatrix}$$

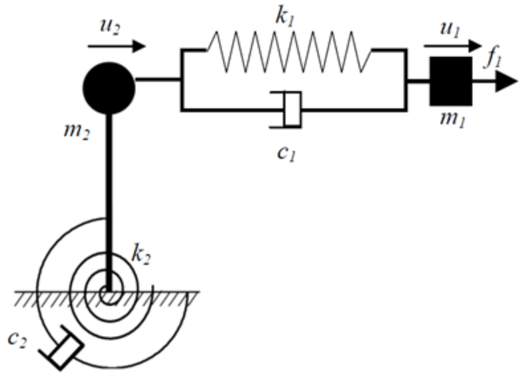

Figure 3.5: In the second-order linear differential equation above, the apparent mass, damping, and intrinsic stiffness of the trunk are denoted by m_2, c_2, and k_2, respectively; and the same for the connecting elements as 1.5 kg, c_1, and k_1, respectively. Also, $\ddot{u}_1$, $\dot{u}_1$, and u_1 represent the acceleration, velocity, and acceleration, respectively, of the connecting elements, as $\ddot{u}_2$, $\dot{u}_2$, and u_2 represent the same kinematics of the trunk.

3.3 Results

Mean (SD) MVEs / MVFs across all three time points (i.e., BE, AE, and AR) of the HPA and LPA groups were similar ($F_{(1,8)} = 0.16$; $P = 0.698$) / ($F_{(1,8)} = 0.15$; $P = 0.709$) with respective values of 888 (266) N / 812 (155) N and 958 (303) N / 855 (188) N. However, there was a significant time ($F_{(2,16)} = 17.06$; $P < 0.01$) / ($F_{(2,16)} = 6.28$; $P = 0.01$) and time-by-group interaction ($F_{(2,7)} = 20.26$; $P < 0.01$) / ($F_{(2,7)} = 5.77$; $P = 0.013$) in measures of MVEs / MVFs (Figure 3.6). MVEs / MVFs of the HPA group decreased ($P = 0.010$) by 43% / ($P = 0.055$) by 22% from BE to AE and then increased ($P = 0.012$) by 33% / ($P = 0.049$) by 14% from AE to AR. MVEs and MVFs for the low exposure group were consistent across all time points.

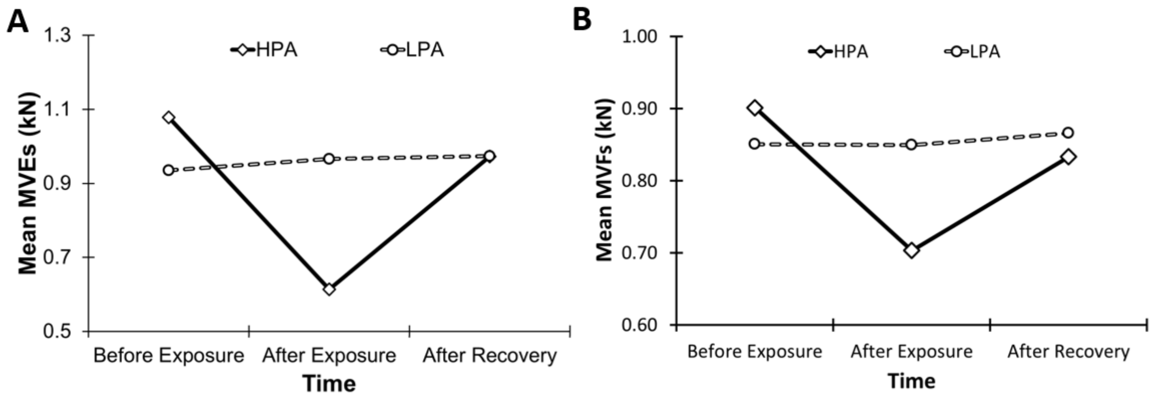

Figure 3.6: Mean values of the (A) maximum voluntary extensions (MVE) and (B) flexions (MVF) for both the high (HPA) and low (LPA) exposure groups.

There was no significant difference ($F_{(1,17)}$ = 0.014; P = 0.9) / ($F_{(1,17)}$ = 0.233; P = 0.636) in predicted values of intrinsic stiffness / apparent mass while maintaining 10% and 30% of MVEs (Fig 3.7), hence statistical analyses were carried out using average values. Such averaged intrinsic stiffness was not significantly different between the groups ($F_{(1,8)}$ = 4.99; P = 0.056) with respective mean (SD) values of 16240 (1918) N/m and 13768 (2144) N/m for HPA and LPA groups. Despite prediction of significant time effects ($F_{(2,7)}$ = 5.86; P = 0.012), none of the post hoc comparisons of intrinsic stiffness demonstrated any significant difference (Table 3.1). However, intrinsic stiffness in the HPA group remained relatively unchanged after exposure but increased 18% (P = 0.054) during the recovery period. Although the HPA group demonstrated a significantly ($F_{(1,8)}$ = 5.41; P = 0.048) higher apparent mass than the LPA group, no significant ($F_{(2,7)}$ = 1.68; P = 0.218) changes in apparent mass occurred with time. Mean (SD) apparent mass (Fig. 3.7D) was 26.4 (3.6) kg / 21.3 (4.0) kg for the HPA / LPA groups.

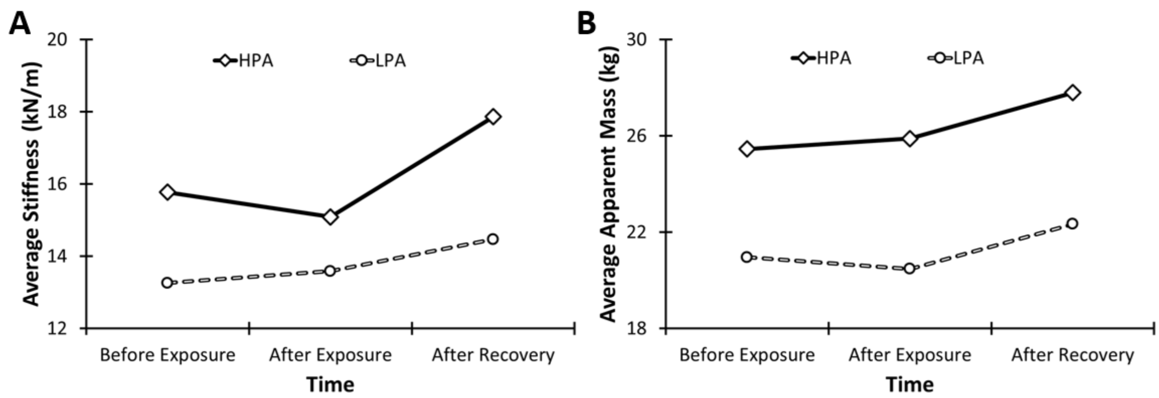

Figure 3.7: (A) / (B) Predicted intrinsic stiffness / apparent mass values averaged across the 10% and 30% efforts for the HPA and LPA groups.

Initial force (IF) / Final force (FF) obtained during the stress-relaxation rests (Figure 3.8A) were similar between groups ($F_{(1,8)}$ = 0.44; *P* = 0.526) / ($F_{(1,8)}$ = 0.71; *P* = 0.424) with mean (SD) values of 216.2 (88) N / 80.5 (32) N and 187.7 (56) N / 89.2 (13.5) N for the HPA and LPA groups respectively. Mean IFs for the HPA/LPA groups increased 28% (*P* = 0.351) / 10% (*P* = 0.219) following exposure and then decreased 28% (*P* = 0.068) / 7% (*P* = 0.209) during the rest period. Corresponding changes in FF were all <30% (P > 0.19).

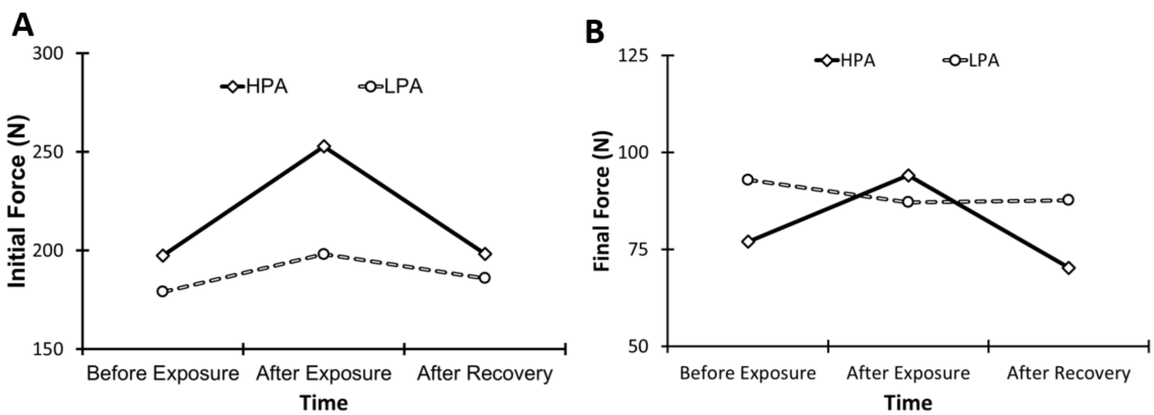

Figure 3.8: Mean (A) initial and (B) final forces during the stress relaxation tests for the high and low exposure groups.

Measurement Differences between Time Points													
		MVE		**MVC**		**K**		**M**		**IF**		**FF**	
		Diff (N)	P	Diff (N)	P	Diff (N/m)	P	Diff (kg)	P	Diff (N)	P	Diff (N)	P
High Exposure	**AE-BE**	-464.7	0.010	-197.9	0.055	-684.0	0.495	0.43	0.907	55.35	0.351	17.10	0.192
	AR-AE	358.0	0.012	129.9	0.049	2777.0	0.054	1.91	0.234	-54.53	0.068	-23.84	0.396
	AR-BE	-106.7	0.399	-68.0	0.285	2093.0	0.061	2.34	0.408	0.82	0.987	-6.74	0.739
Low Exposure	**AE-BE**	30.4	0.445	-1.0	0.977	328.7	0.732	-0.48	0.698	19.05	0.219	-5.75	0.536
	AR-AE	8.1	0.820	16.4	0.491	879.7	0.119	1.86	0.004	-12.18	0.209	0.50	0.949
	AR-BE	38.5	0.259	15.3	0.698	1208.5	0.229	1.38	0.317	6.87	0.624	-5.25	0.650

Table 3.1: Differences between trials and their significance.
(Significant values are highlighted.)
(MVE = Maximum Voluntary Extension, MVF = Maximum Voluntary Flexion,
K = Intrinsic Stiffness, M = Apparent Mass, IF = Initial Force, FF = Final Force,
BE = Before Exposure, AE = After Exposure, AR = After Recovery)

3.4 Discussion

The primary goal of this study was to investigate the effects of level of exposure to physical activity on diurnal changes and overnight recovery in aspects of lower back mechanics. It was hypothesized that exposure to a higher level of physical activity will be associated with greater alterations in aspects of lower back mechanics and will not fully recover following an overnight rest period. Changes in measures of maximum voluntary exertions (i.e., MVEs and MVFs), as expected, were greater among the individuals in the HPA group as compared to those in the LPA group. However, contrary to our hypothesis, changes in MVE and MVF efforts of the HPA group completely recovered following the overnight rest period. Predicted higher apparent mass of the HPA group than the LPA group was consistent with the higher average weight of the HPA group members as compared to those in the LPA group. Apparent mass was not affected by any diurnal changes in active and passive aspects of lower back mechanics.

Since perturbation tests were conducted in an upright trunk posture (i.e., a posture with minimal passive stiffness), changes in intrinsic stiffness were mainly affected by alterations in the level of exertion prior to perturbations. Hence, it was expected that changes in trunk intrinsic stiffness among the HPA group members to follow the same trend as the changes in MVEs. However, prediction of no changes in intrinsic stiffness among members of the HPA group, despite a significant decrease in their MVEs, suggests an altered trunk neuromuscular patter which could keep intrinsic stiffness at the same level as before exposure. Though not significant, the increase in intrinsic stiffness of the HPA group following the overnight recovery suggests continuation of the altered trunk neuromuscular behavior despite complete recovery of MVE.

We did not find any significant change in measures representing the passive aspects of lower back mechanics. Creep deformation of the spine has been shown to recover substantially following an hour of recovery period (Bazrgari et al., 2011). It is likely that the rest periods during the day along with the time elapsed between the final tournament game and the start of the stress-relaxation test was sufficient for recovery of the passive aspects.

Although there are several limitations associated with the present study, results suggest a potential for accumulation of diurnal changes in lower back mechanics following a day with high level of physical activities. The main limitation of our study was the small sample size which could likely be a reason for our failure to observe significant change in some of the measured aspects of lower back mechanics. Nonetheless, current results motivate future studies with a reasonable sample size. We conducted all of the measurements in the sagittal plane, hence any changes in lower back mechanics in the other planes were undetected. Finally, a crossover study design wherein study participants undergo both exposure levels would have been a more suitable study design to test our hypothesis.

3.5 Conclusions

Both MVE and MVF measurements of the HPA group followed the expected path by decreasing following fatigue and returning following recovery. Conversely, the intrinsic stiffness remained the same following a day-long intensive physical activity and then increased following recovery. This may be a result of alternate trunk neuromuscular behavior to maintain a minimum threshold of trunk stiffness to prevent injury and keep spinal equilibrium and stability.

Chapter 4: Post-Experiment Improvements

Based on the performance of the perturbation system following several months of data collection, several modifications were made to different parts of the system, including the harness, the leg platform, and the motor platform.

The harness was proven to be restrictive in the abdominal region causing minor discomfort for the participants. Ergo the bottom halves of both the chest plate and the back plate of the harness were cut off so that the edges were near the bottom of the participant's sternum. This proved advantageous when the new experimental procedures called for placing EMG electrodes on the abdomen. This also fixed the problem of the harness compressing during MVF effort trials. During these flexion efforts, when the participant pushed passed a certain threshold of force, the harness would cave inward allowing the participant to fall forward about a millimeter. This caused the participant to lose his/her sense of the required pushing force during the perturbations. Since the harness no longer makes contact with the participant's abdomen, it can be clamped tighter thus removing the cave-in phenomenon. Since the harness was halved, the u channels on both sides needed to be raised. This also improved the angle of both the connecting cylinder on the front of the harness and the laser sensor plate on the rear. The connecting cylinder proved to be less durable than anticipated; this could be due to its soft aluminum material. Therefore it was replaced with a ½" steel bolt, with the same ¼" hole cut through it. Both of the plates being used to mark displacements for the laser sensors, the one in the connecting rod and the one on the rear of the harness, were replaced with sturdy aluminum plates.

Due to changes in the experimental procedure that requires participants to be lower in the leg platform and wear EMG electrodes, the leg platform needed to be improved. Since the participants were lowered, the seat was too high to be effective and controlling the rotation of the pelvis. Therefore the seat and its frame were removed. In replacement of the seat, an extra (7” x 29”) of the wood and plastic boards were added as extensions of the current boards. An aluminum profile (29” x 1” x 1”) was placed horizontally above these extensions and bolted to the back side of the vertical profiles. This profile served as the new attachment location for the seat belt. As a result of these changes, the EMG electrodes on the participant’s erector spinae are no longer compromised by the seat, and the pelvis rotation seems to be better controlled during the stress relaxation tests. Also as a result of the changes to the leg platform, the motor platform needed to be lowered significantly in order to maintain the same spectrum of eligible participants.

Chapter 5: The Wobble Chair

5.1 Introduction

The perturbation system provided a platform for evaluating the mechanical contributions of passive and active trunk tissues to lower back mechanics (i.e., mass, damping, stiffness, passive resistance, etc.). However, another system was needed in the HMBL to further understand the relationship between the trunk stability, reflex response, proprioception, and the neuromuscular system.

Hemisphere-shaped balance trainers have been used by many physical therapists and gyms as a way to strengthen the core, improve trunk stability, and decrease some cases of low back pain. During balance trials continuous kinetic and kinematic variances cause small biomechanical and neuromotor disturbances to the neuromuscular system. The neuromuscular system uses sensory orientation cues (proprioception) and external demands to control the reflexive and voluntary muscle responses in order to maintain an upright, zero-velocity posture. Wobble chairs (See Figure 5.1) use a similar concept to test the user's balance and stability but provide a measureable outcome of the user's performance. Wobble chairs consist of a seat resting on an unstable support that allows tilting but prevents translational movement. Most wobble chairs also have additional supports with variable resistance to provide assistance to the user as he/she attempts to maintain an upright posture. Several different numerical methods have been developed to assess stability on the basis of center of pressure data (B. Hendershot, 2012; Lee, 2007; Tanaka, 2008).

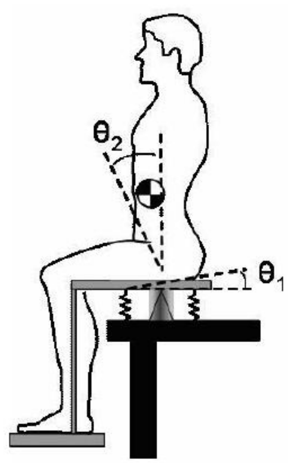

Figure 5.1 (Lee, 2007): Participants on the wobble chair must control the angle of the seat (θ_1) by altering their trunk angle (θ_2).

5.2 Model Designs

The wobble chair at Virginia Tech (B. Hendershot, 2012; Lee, 2007; Tanaka, 2008) served as the model for our design (Figure 5.2). Briefly, this wobble chair consists of three main components: the seat, the spring system, and the base. The seat, adjustable in the anterior-posterior direction, uses a pelvic strap and an adjustable foot rest to keep the user's lower body firmly attached to the chair. It lays on top of a low-friction ball-and-socket joint and adjustable spring system. The springs can be moved inward to decrease the stability of the chair. The spring system is rigidly attached to a force platform at the top of the metal base. The force platform measures the load response of the user as he/she attempts to maintain an upright posture while sitting on it (B. Hendershot, 2012). Seat and torso angles in both the anterior-posterior and medial-lateral planes were measured using 6 degree of freedom electromagnetic motion sensors (Tanaka, 2008).

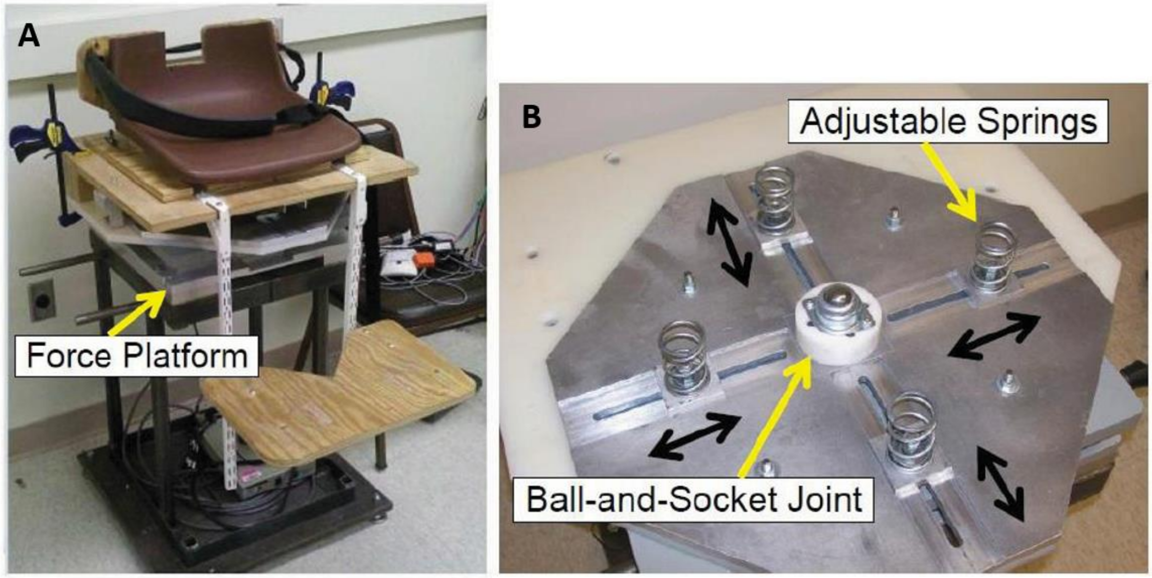

Figure 5.2 (B. Hendershot, 2012): The seat of the wobble chair lays on top of the ball-and-socket joint providing instability. The springs provide an adjustable amount of assistance to the participant's objective of keeping the seat level.

Since this wobble chair was recently developed, the main experiments performed using it have been pilot studies. This includes studies to find the optimal test duration to process stationary of torso stability assessment, the intra-session and inter-session reliability of the stability measures (Lee, 2007), and the effects of testing with the participant's eyes open and closed (Tanaka, 2008). However, there have been other studies using similar systems to test the effects of horizontal forces on the torso (Lee, 2007) and the effects of lower-limb amputation (B. Hendershot, 2012) on trunk postural control and stability.

Prior to the construction of our own wobble chair. A senior design team at the University of Kentucky was tasked with creating a low cost wobble chair prototype capable of tracking the user's center of gravity by methods other than a force plate. Their design, as shown in Figure 5.3, was loosely based on the design from Virginia Tech and utilizes cantilever strain gauges attached to the base of location-adjustable springs. The prototype design was essential for the development of the actual wobble chair. In particular, from this initial design we learned that the height between the seat and the base is critical for the safety of the user. In the case of excessive height between the seat and the base, there is a high risk for the seat to fall off the ball joint.

Figure 5.3: The prototype design uses strain gauges instead of a force plate to track the user's movements.

5.3 HMBL Design

The wobble chair (see Figure 5.4) was constructed similarly to the design at Virginia Tech. The components of the wobble chair can be categorized into four groups: the frame, the force plate, the spring system, and the seat system.

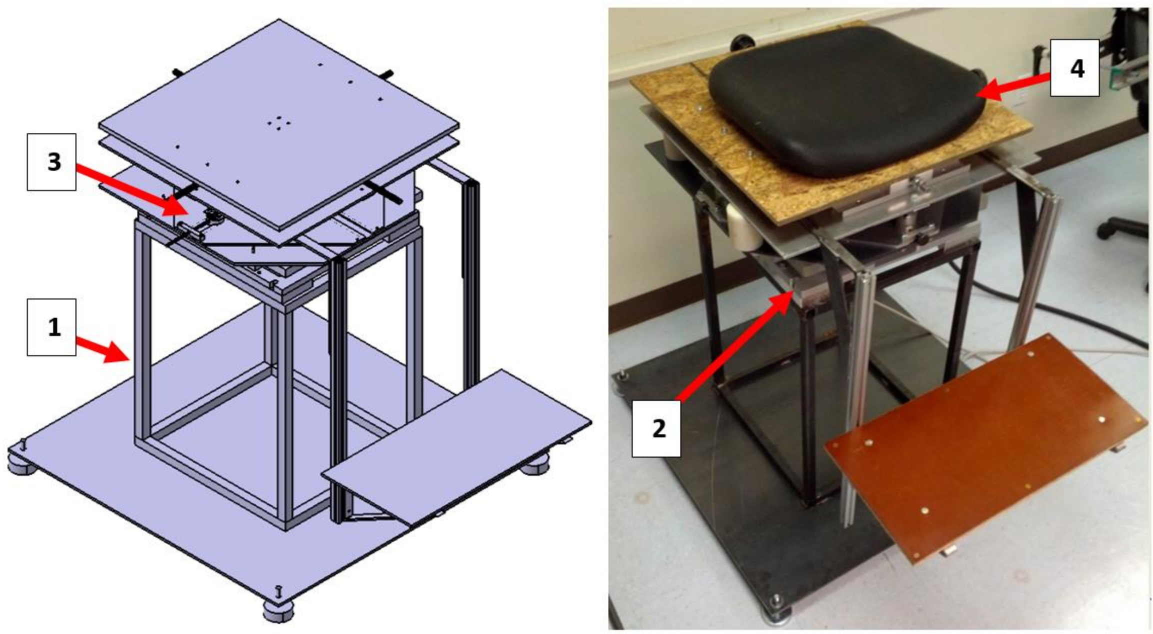

Figure 5.4: Wobble chair in the Human Musculoskeletal Biomechanics Lab at the University of Kentucky. Four groups of components: 1) the frame, 2) the force plate, 3) the spring system, and 4) the seat system.

A 3' x 3' x ¼" hot-rolled steel plate was implemented at the base of the frame (see Figure 5.5) to enhance the overall stability of the system, thus preventing the entire wobble chair from tipping over in any direction while the participant is using, entering, or exiting the chair. A ½"-13 tapped-hole was drilled at each corner, 1" from each side, where level mounts (McMaster-Carr, Aurora, OH) were attached to keep the base plate off the floor. The remainder of the frame is made of 1" x 1" x ¼" steel square tubes. These tubes were welded into a 20" x 18.1875" x 24" rectangular prism. The 20" x 18.1875" side of the prism was then welded to the center of the base plate. Before the top of the prism was welded together, four ¼" holes were drilled in it to attach the mounting rails of the force plate. Since nearly the entire frame is made of steel, the base of the wobble chair is very rigid, increasing safety for the participants and preventing any errors due to vibrations.

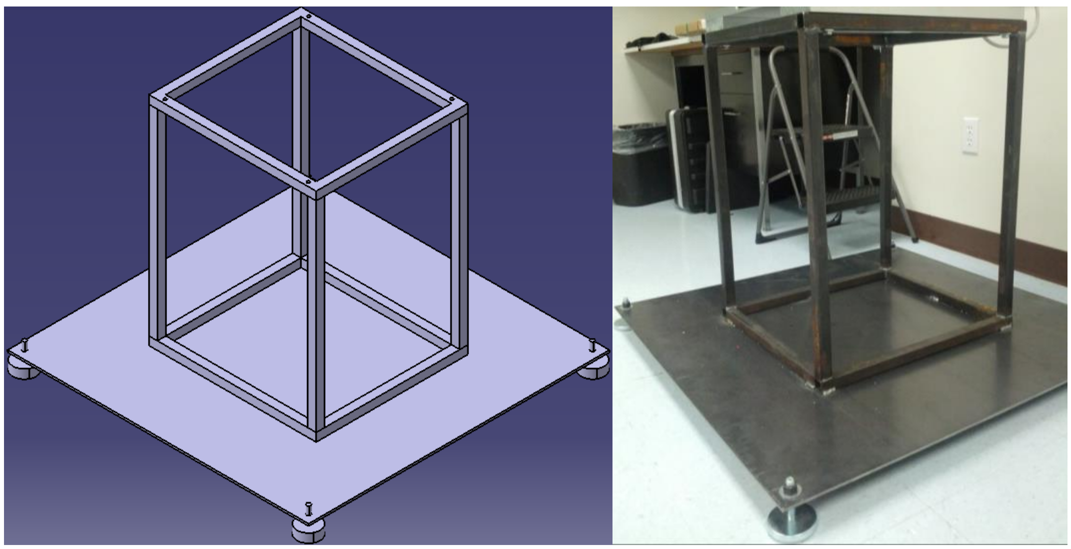

Figure 5.5: The frame consists of square steel tubes welded into a rectangular prism and onto a steel plate.

The force plate (AMTI, Watertown, MA) was packaged with two mounting rails that were easily bolted to the top of the steel rectangular frame (see Figure 5.6), thus facilitating the attachment of the force plate to the frame. Instead of drilling the upper surface of the force plate for the attachment of the spring system, an attachment mechanism was designed to tightly clamp the spring system on the force plate. Provision of such attachment design was to allow application of the force plate in other studies when it is not in use inside the wobble chair. The attachment mechanism includes a 2' x 2' x ¼" plastic plate and six sets of aluminum blocks and bolts as braces. Each set of braces consisted of a 1” x 2” x 5/8” horizontal and a 1” x 15/16” x 5/8” vertical block with a ¼” hole through them to allow for a bolt to tighten them together on the rim of the force plate making it a removable and non-damaging option for attachment to the plate. The plastic plate also has four holes in its center to attach to the spring system that sits on top of it.

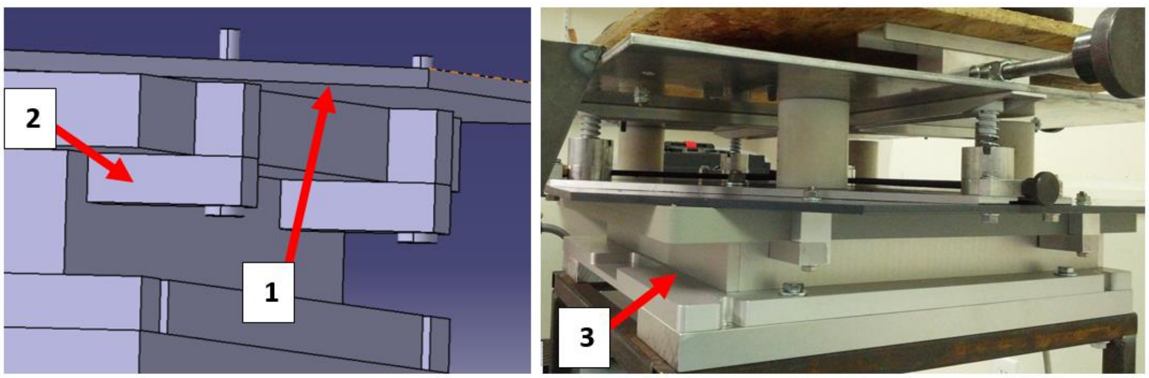

Figure 5.6: The (1) plastic plate is attached to the (3) force plate via the (2) braces.

The spring system (see Figure 5.7) was designed to provide a variable degree of support for the seat system. This was achieved by supporting the seat system using a set of four springs with adjustable distances from the center of the spring system where a ball joint connected the spring system to the seat system. By moving the springs toward the center of the system (i.e., toward the ball joint), the seat system would become less stable. The base of the spring system was made from a 2’ x 2’ x ¼” aluminum plate. Two grooves were milled on top of each other in the plate to create four lanes for the springs to travel. Each groove was milled 1/8” deep and 10.5” long, but one was 1.625” wide and the other 7/8” wide. These grooves allow the spring bases to slide in only one direction without any rotation (See Figure 5.7.3). The spring bases were made from a 1.625” diameter aluminum cylinder and cut 3/4" thick. Two slots were cut into each spring base

1/8” tall and 3/8” deep, 1/8” above the bottom. A ¼”-20 tapped-hole was drilled into the center base, 1/4" from the top, in the same direction as the slots. This hole will be used to adjust the springs towards and away from the center of the plate via threaded rods, thus changing the degree of difficulty for the participant. The base plate has sixteen bolt holes used to attach three groups of the spring system: a) four 7/32” holes in the center, used to attach the ball transfer base; b) eight ¼”-20 tapped-holes on the outside, used to attach the threaded rod guides; c) four 3/8”-24 tapped-holes in the middle, used to attach the underneath plastic plate. We used four 2” zinc-plated steel compression springs (McMaster-Carr, Aurora, OH) for the spring system. Each spring was placed inside a ¾” hole in the spring base, and had a plastic T on its free end to facilitate sliding along the aluminum channels under the seat system. At the center of the spring system sits the base of the ball transfer which was created from a 3” aluminum cylinder and cut ¾” thick. The base of the ball transfer provides an attachment side for the threaded rods that move each of the springs. The attachments are four ¼” holes that were cut into the side of the ball transfer base, ¼” from the bottom and 90^{o} apart. The ball transfer (McMaster-Carr, Aurora, OH) and its base are connected to the base plate using four ¼”-20 bolts. The adjustability of the springs is provided by four guiding threaded rods (1/2” x ½” x 3.125”) that at one end are connected to the base of the ball transfer and at the other end have a steel knurled-rim knob (McMaster-Carr, Aurora, OH). Each rod runs through the base of a spring and a bearing system on the edge of the spring system’s plate. A pair of nuts were added to each guiding threaded rod on both sides of the guides to prevent the rod from moving axially. A ¼”-20 tapped-hole was drilled into the ball within the ball transfer and attached to the ball transfer cap, connecting the ball transfer to the seat system above it.

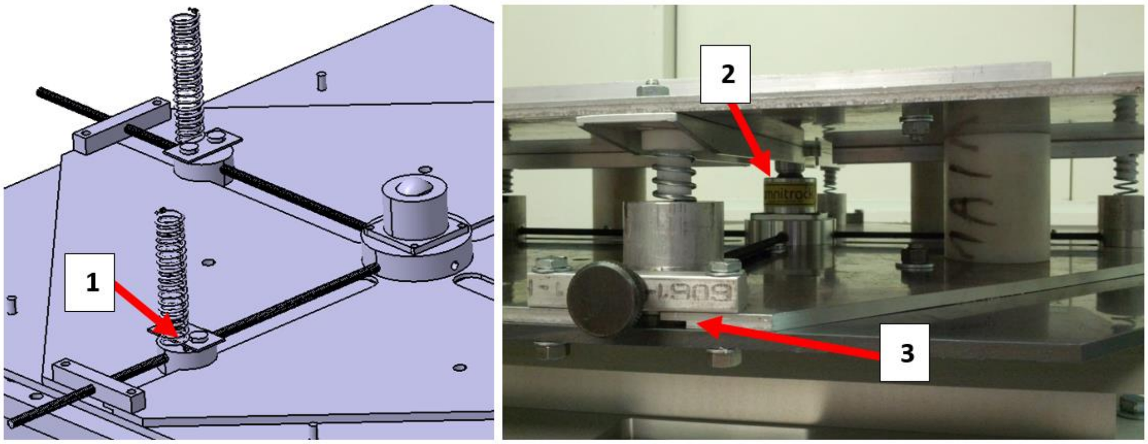

Figure 5.7: The (3) offset grooves allow the (1) spring bases to move only towards and away from the (2) ball transfer.

One important improvement in our design was due to the need for seat adjustability. Since each user will have a different center of mass, the seat needs to be adjustable in both directions parallel to the seat so that each user will need to maintain the same upright posture for equilibrium. The seat system (See Figure 5.8) was designed to accommodate such adjustability of the center of mass. This was done by designing a two degrees of freedom sliding system using two parallel plates. One of the plates (2’ x 2’ x ¼” aluminum) sits on the spring system and the other one (2’ x 2’ x ½” wood board) is bolted to the seat. Attached to the underside of the seat plate is a 4” x 1.5” x 1.25” aluminum cube and two 1.5” x ½” x 1/8” aluminum u channels (Figure 5.8.2). An identical set, rotated 90^{o}, is attached to the topside of the other plate. The aluminum cubes have a ½”-13 tapped-hole through the sides opposite the u channels. These two plates are separated from each other by four aluminum blocks (Figure 5.8.1). Each aluminum block has a 0.5” through-hole, 1” above/below the surface contacting the u channel. Two ½”-13 threaded rods, one for each direction, are placed through the center cubes, and a pair of the four sliders. Two nuts are placed on both sides of each slider along each threaded rod, and a steel knurled-rim knob (McMaster-Carr, Aurora, OH) is placed on one end. Therefore, as each threaded rod is rotated, the seat plate moves laterally causing the sliders to slide along the u channels.

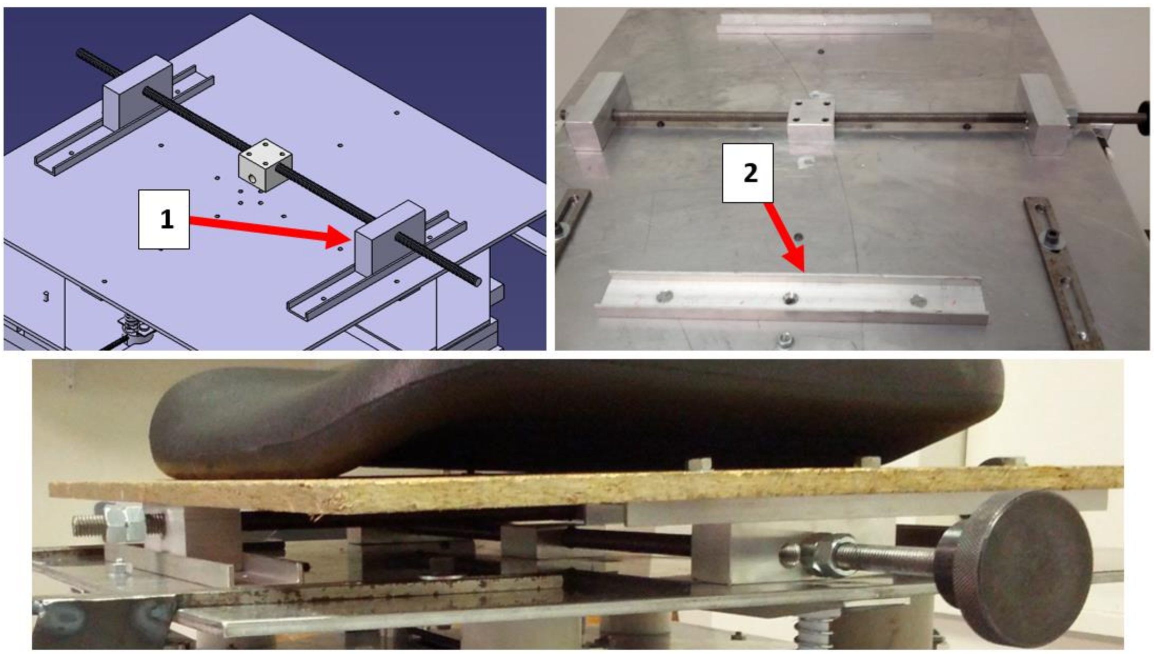

Figure 5.8: Two pairs of (1) aluminum blocks slide along the (2) U-channels to give the seat two dimensions of adjustment.

The main improvement made on this wobble chair, in comparison with the design at Virginia Tech, is the ability to adjust the seat not only frontwards and backwards but also side to side. In a previous study performed at Virginia Tech (B. Hendershot, 2012), persons with unilateral lower-limb amputations performed stability tests on the wobble chair. Although the study did not divulge how the medio-lateral center of mass offset was corrected, the two-axis adjustable seat would provide a quick and efficient solution.

Chapter 6: Summary and Conclusions

This project was designed and implemented in order to gain a better understanding of trunk mechanical and neuromuscular behaviors aiming at the control and management of low back pain. To achieve this purpose two systems were developed for use at the Human Musculoskeletal Biomechanics Lab at the University of Kentucky: 1) a displacement controlled sudden perturbation system, 2) a wobble chair system.

A thorough review of literature on the current perturbation designs that are used to assess lower back mechanics proliferated the criteria for the HMBL perturbation system. This includes the abilities to estimate changes in the passive, active voluntary, and active reflexive aspects of trunk mechanics. The perturbation system consists of six main subsystems: 1) a rigid frame capable of absorbing the vibrations caused during perturbation and providing a structure for the other subsystems, 2) the adjustable leg platform for restraining the user's pelvis to the system and rotating the user's legs during stress relaxation tests, 3) the motor platform providing adjustable yet sturdy housing for the motor which controls the trunk's motion during perturbations, 4) the harness facilitating a firm connection with the user's trunk while maintaining comfortability, 5) the adjustable connecting elements containing the load cell and target for the laser displacement sensor, and the electrical components controlling the system including the safety features.

This system was then constructed and tested with a pilot study. This study sought to quantify the effects of level of physical activity on diurnal changes and overnight recoveries in aspects of lower back mechanics. Two groups were exposed to high or low levels of physical activity and tested three times: 1) in the morning before exposure, 2) in the evening following exposure, and 3) in the following morning during recovery. Each session consisted of three tests: 1) standing maximum voluntary exertions, 2) displacement controlled perturbations, and 3) stress relaxations. Both MVE and MVF measurements of the HPA group decreased following fatigue and returned following recovery. The intrinsic stiffness remained the unchanged following a day-long intensive physical activity and then increased during recovery. This may be a result of alternate trunk neuromuscular behavior to maintain a minimum threshold of trunk stiffness to prevent injury and keep spinal equilibrium and stability. All other data trends proved to be insignificant due to the small sample size in this study. Following this study, modifications were made on the perturbation system to improve its performance.

The project concluded with the design and assembly of the wobble chair. It consists of 4 major subsystems: 1) the frame providing a vibration-resistant structural support, 2) the force plate measuring the user's forces and moments, 3) the spring system offering adjustable resistance to the user allowing variable difficulty, and 4) the seat system providing the adjustments necessary to center the user on top of the system. The outfitting the HMBL with the perturbation system and wobble chair is hoped to facilitate future research aimed at a better understanding of trunk mechanical and neuromuscular behaviors to prevent and treat LBP in the future.

Appendix

Trunk Movements during Ultimate Frisbee

Athletes who play Ultimate Frisbee (often shortened to ultimate) were ideal candidates for the pilot study presented in chapter 3 because of the amount of physical demand exposed on their trunk during the game. Although no research has been performed on the prevalence of low back pain in ultimate athletes, personal experience has shown that it is common for many players to have sore trunk muscles after tournaments as well as trunk injuries. In fact two of the players scheduled to participate in the chapter 3 study could not due to trunk injuries developed during the tournament. During the tournament the athletes participated in four games per day (lasting about 1.5 hours per game) wherein they experienced a variety of back movements throughout each game.

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