A NEW METHOD TO ESTIMATE THE IMPACT ON THE L5/SI SPINAL DISC FROM SPEED LIFTING OF UNSTABLE LOADS, ASYMMETRICALLY

by

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Abstract

Background: The most significant causes of lower back injuries at work are probably from manual lifting activities. Lifting unstable loads pose a significant strain to the lower back and can cause debilitating lumbar spine injuries. The primary function of the vertebral column is to support the upper body. The L5/S1 disc junction located between the lumbar and the sacral regions of the vertebral column is the most critical joint in spine with respect to lifting strain. Because of its position and the amount of upper body weight it handles, it is particularly vulnerable to misalignment, wear and tear, and injury. Lifting loads has several adverse effects on the back: spinal compression and excessive strain on the back's tendons and ligaments may result in herniated or ruptured discs, disc degenerations, spinal stenosis, and other diseases. Biomechanical models have been developed to estimate the forces in the lumbar spine to determine the severity of lifting tasks, and NIOSH has developed an equation to determine a safe lifting load weight. However, neither covers all important types of lifting conditions adequately. The NIOSH equation does not include the lifting load type (stable versus unstable loads) as a variable due to insufficient research information, and the measurement of spinal force and body motion variables still need improvements to adequately measure and assess the severity of lifting unstable loads asymmetrically.

Objective: The main objective of this study is to demonstrate a new method to measure compression, shear, and torque forces in L5/S1 disc when lifting unstable loads under different lifting conditions (load type, lifting style, and lifting speed).

Methods: A total of 3 subjects volunteered for this study to lift a bin, partly filled with water, with various lifting conditions while standing on force plates. Each subject lifted the bin (stable and unstable load) at two different speeds (normal and fast), and along two different planes of the body (sagittal and frontal) – a total of 8 lifting conditions. The water induced the required instability in the load during lifting. For simulating stable loads (of the same weight – 30 lb) solids replaced water in the same bin. Body joint angles, velocities, accelerations (using a visual 3D Vicon camera system), and forces from force plates and load cells along with 12 electromyography (EMG) signals for detecting muscle contractions were used to determine compression, shear and lateral forces on the L5/S1 disc. To get a more precise force impact on the L5/S1 disc, a technical coordinate system was created to define the orientation of the L5/S1 disc. A bottom-up approach of Newton-Euler dynamics was used using visual 3D software to estimate compression, shear, and torque force on estimated virtual landmark of L5/S1 disc.

Results: This study has provided an alternative method to measure compression, shear, and torque forces at estimated location of L5/S1 disc proximal to the technical virtual landmarks. The results have shown to be consistent with other published research papers.

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Chapter 1: Introduction

1.1 Background

On a daily basis, workers are required to interact with a variety of equipment and tools to accomplish various tasks. Some of these tasks require manual material handling (MMH) and involve lifting, lowering, holding, carrying, pushing or pulling activities. In some cases, these tasks require manipulation of heavy loads. These types of MMH activities are physically stressful in the workplace and are associated with numerous cases of injuries and Cumulative Trauma Disorders (CTD).

1.2 Etiology of low back injury

Epidemiological, biomechanical, and physiological research provides the necessary insight into the causes of musculoskeletal disorders. Research has shown that lifting tasks have been the highest contributor to, or cause of, lower back injuries, accounting for 49-60% of lower back incidents (Eastman, Kodak, 2nd ed., 1986). According to the U.S. Bureau of Labor Statistics (BLS Report, 2009), musculoskeletal disorders (MSD) accounted for 28% of all work related injuries, and the back was injured in nearly half of these cases, with an average of over 7 days to recover. MMH activities expose workers to physical conditions that are directly related to workplace exposures leading to musculoskeletal disorders of upper extremities and to lower-back and neck injuries (Bernard et. al, 1997).

Low back pain (LBP) is one of the most common health problems in the workplace, affecting the population indiscriminately (BLS Report, 2009). The Occupational Safety and Health Administration (OSHA) reported that there are over 1 million workers a year with low back pain. It is estimated that 8% of the entire working population might be disabled per year, accounting for 40% of all lost working hours. Employers lose useful manpower hours and may need to employ, train, and pay temporary replacement workers a hefty price, costing up to 100 billion U.S. dollars per year in lost time, wages, and health insurance (Klein et al., 1984; Vojtecky et. al., 1987). OSHA (1990) reported that the claim costs of back pain were more than double the average cost of other types of compensation claims. Over 60% of lower back injuries are caused by over exertions in the workplace, the second highest cause of days off after the common cold and flu (Klein et al., 1984; Vojtecky et. al., 1987). The National Institute of Occupational Safety and Health (NIOSH, 1981) reported that 60% of people who suffered from lower back pain experienced overexertion from their daily jobs and less than one third of employees returned to their jobs after a lower back injury. Some reports claim that over half of the occupational injuries are due to job over-exertions and approximately 60% of overexertion injuries involve lifting loads, while 20% are due to pushing and pulling (NIOSH, 1981). Lewis and Narayan (1993), and Kar et al., (2003) have shown that job overexertion contributes to decreased productivity and more discomfort, and it may also result in increased biomechanical stresses, fatigue, injuries, accidents, and cumulative traumas. Workplace tasks still involve many MMH activities, despite mechanization, since machines are often unnecessary, unavailable, costly, or not appropriate for the job.

1.3 Etiology of MMH

MMH injuries are typically associated with exertions that involve one or more of the following stresses: repeated or constant exertions, deviated (awkward) postures, vibrations, or high contact forces (Chaffin et al., 2006). The National Institute of Occupational Safety & Health (NIOSH, 1981) stated that the most problematic issues with MMH affecting the human body were improper methods of lifting, pulling, pushing, carrying, lowering, bending, twisting, unexpected exertions, slipping, or falling. Many manual material handling (MMH) tasks are being replaced by machines and have undoubtedly decreased the number of workers required to perform strenuous manual handling jobs. However, it is not clear how quickly the displacement of strenuous tasks occurs. MMH will continue to exist since many manual tasks are not likely to be automated or mechanized, particularly in the growing service industries, for example, construction, mechanical repair, baggage and package handling, police protection, military services, medical services, etc. Chaffin et al. (2006) and Waters et al. (1994) indicated that MMH activities will continue to be prevalent in many industries and will likely be the main contributors to musculoskeletal disorders in the U.S. workforce in large numbers.

1.4 Lifting Posture

Posture: One of the most troublesome factors in MMH injuries is improper or poor work postures. Postures that result from bending, twisting, and overreaching are among the worst that contribute to musculoskeletal disorders (MSD) in industrial work (Madeleine et al., 1998; William et al. 2000). The Bureau of Labor Statistics Report (2008) stated that occupational tasks that require workers to work at or below knee-height (such as landscaping, roofing or concrete work) for prolonged periods of time have relatively high incidence rates of low back disorders (LBD) compared to the industry average. Corlett et al. (1986) and Kilbom (1988) have reviewed the relationship between working posture and industrial efficiency and health. They found that occupations that require tasks involving extended periods of stooping (e.g. farming, etc.) rank as the most problematic with respect to work-related LBDs (Goldsheyder et al., 2002; Rogers and Granata, 2006; Adams and Dolan, 1996). These constrained postures often require muscular efforts and body motions that some workers are incapable of maintaining throughout a regular workday. Performing jobs in prolonged standing postures alone can be linked to lower back pain among industrial workers, due to reduced blood circulation in the lower legs, and localized muscle fatigue. Additionally, continuous standing for more than 4 hours a day is likely to contribute to pain in the lower back and feet (Lafond et al., 2009, Madeleine, 1998, William et al., 2000 and Messing and Kilbom, 2001).

1.5 Mechanics of Lifting Loads and Trunk Stability

It is necessary to understand the mechanics of load lifting in order to understand the interplay of the muscle and joint forces and movements of joints in the body. According to Schultz and Andersson (1981), lifting is the organized collaboration of different body parts in a critically timed system, orchestrated to organize in a proper and timely manner to achieve one's task. At each phase during lifting, a lifter may utilize different postures, constrained by a number of situational and personal variables. During each phase of a lift, the lumbar spine will absorb the major portion of the forces -- any external loading (including the load to be lifted), the upper body weight, muscular contraction, restorative forces of passive ligaments, and any forces secondary to intra-abdominal pressure (IAP) (Schultz and Andersson, 1981; Schultz et al., 1982a, b). The biomechanical reactions of the body are neither simple nor uniform. A wide range of muscle activity changes and spinal loading characteristics may occur for different tasks (McGill et al., 2004; Marras et al., 1998), and spinal loading may even change from one exertion to the next, for identical tasks.

Load lifting, in general, may have several negative effects on the musculoskeletal system of the body, including spinal compression and excessive strain on the tendons and ligaments. The lumbar back system is disturbed during lifting of a standard stable load, from the start to the destination of the lift (Anderson et al., 1986; Chaffin, 1979; McGill and Norman, 1985; Schultz and Anderson, 1981). This is especially true for workers who handle shipments that deal with millions of loads and different types of products, such as solids, liquids, gases, etc., that are being transported from one place to the other.

Lifting of loads involves stable and unstable lifting load types. Past research has indicated that beverage manufacturing industries have accounted for 60% of lumbar spine sprains and strains (McGlothlin, 1996), and that the lifting of unstable loads (liquid beverages) was the most common factor among all the beverage industries.

1.5 Purpose and objectives of the study

Purpose and Objectives: The main purpose of this study was to demonstrate a new method to measure compression, shear, and torque forces in the spinal L5/S1 disc when lifting unstable loads (with a shifting center of mass), asymmetrically, and at different speeds, simulating typical work methods of lifting such a load. This study captures abdominal and erector spinae muscle contractions, ground reaction and load cell dynamic forces, along with kinetic (spinal forces) and kinematic variables (Marras et. al, 1998; Schipplein et al., 1990) to estimate compression and shear forces on the L5/S1 disc, and torques about the disc.

The specific objectives in this methodology are to:

i. Design a customized bin with load cells to be used for estimating X, Y, Z forces in the human lumbar spine, as the unstable center of mass of the load (liquid water) shifts during lifting.

ii. Create a technical coordinate system using the Vicon software to define a more precise location of the L5/S1 disc.

iii. Build a comprehensive method that can sum all the forces impacting the L5/S1disc during the lifting of the unstable loads.

iv. Measure and analyze compression and shear forces, and torques, generated at the L5/S1 disc during lifting of the unstable load.

1.6 Novelty of Approach

The following aspects delineate originality of this research:

- Estimation of compression, shear, and lateral forces on the L5/S1 disc, using kinematic variables and EMG data while lifting unstable loads, at fast speeds, with a 90-degree asymmetrical twisting of the upper body.
- Adopting the use of EMG and kinematic variables to provide a cross validation of the force impact on L5/S1 disc.
- Designing a technical coordinate system to transform L5/S1 disc to a more precise location specifically within vertebras L5 and S1.

1.7 Contribution to Knowledge

Manual lifting of unstable loads is a common daily task required by many of workers. The methods of measurement may reveal unknown risks to lower back injuries and could be used to develop improved lifting techniques when dealing with unstable loads, relating to lifting speed or symmetry. This study may also help in the design of future research in lifting studies.

Chapter 2: Literature Review

2.1 Structure of the Lumbar Spine

The vertebrae of the human spine are separated from each other by intervertebral discs, keeping the bones from rubbing together. The discs act as shock absorbers or cushions to forces created from physical tasks, motion and sustained postures, and permit movement of the vertebrae relative to each other. Anatomically, the discs consist of two parts, the inner nucleus pulposus and the surrounding annulus fibrosus. The nucleus pulposus is a gelatinous semi fluid responsible for much flexibility and resilience of the vertebral column. Forward flexion, lateral flexion, extension, and rotation are the only movements of the lumbar spine to facilitate supporting the upper body (Kumar and Chaffin et.al, 1988). However, these movements result in changes in the compressive, shear, and torque forces on the lumbar spine (Adams, 2002). Compression forces within the nucleus pulposus increase and push on the annulus fibrosus when the spine flexes (Chaffin, et al., 2006). Shear forces occur when one vertebra moves forward, backwards or sideways with respect to its adjacent vertebrae. Work related to overexertion on intervertebral discs with repetitive compressive forces may cause the annulus fibrosus to tear, leaking the internal nucleus pulposus, and resulting in what is called disc herniation (Chaffin, et al., 2006). Herniated discs occur more frequently with repetitive extreme forward bending of the spine in MMH workers (Adams and Hutton, 1982).

The majority of lower back injuries are believed to be caused by overloading the back-extensor muscles and spinal tissues during MMH lifting tasks. These include muscular and ligamentous injuries; degenerative changes in intervertebral discs;

herniation of the intervertebral discs with irritation of adjacent nerve roots; or degenerative changes in the intervertebral discs (Deyo, Rainville, and Kent, 1992; Trzcielinski et al., 2012).

2.1.1 Compression Forces on L5/S1 Disc

The lumbosacral joint (L5/S1 disc) is the most critical joint in relation to low back pain (LBP) because of its position in the spine and the amount of upper body weight it bears. A considerable amount of force is generated by the back-extensor muscles on the vertebral discs L5/S1 when lifting weights (Kumar and Chaffin et.al, 1988). NIOSH (1991) has considered the L5/S1 disc compressive force as the only critical biomechanical stress factor for determining the risk to low back injury (Norman et al., 1998); and quantifying this force, is therefore critical in assessing the severity of the causative or associative biomechanical stress. The two main methods for determining the L5/S1 disc compression are: (i) quantifying all the forces generated by the muscles attached at the L5/S1 joint (Cholewicki et al., 1995; Marras & Granata, 1997; Fathallah et al., 1998), and (ii) and analyzing the net forces on the L5/S1 disc (compressive, shear and joint moment forces) (Buseck et al., 1988; Bush-Joseph et al., 1988; Cholewicki et al., 1995; Kingma & van Dieen, 2004a). Early research on this topic utilized summative estimation of compressive and shear forces on the L5/S1 joint, calculating the attached muscular and ligament forces on the joint, estimating contraction and co-contraction moments, or estimating the intra-abdominal pressure (Fogleman and Smith, 1995; Mital, 1984; Snook and Ciriello, 1991; Marras et al., 1984; Nachemson, 1981 and Wilke et al., 2001). According to NIOSH, in the biomechanical approach, a compression value of

L5/S1 disc above 3400N would be considered potentially hazardous for workers regardless of gender or age (Chaffin et al., 2006; Water et al., 1993), and a compression force over 6400N, dangerous.

2.3 Role of Muscles

The understanding of spinal stability requires understanding the role of trunk muscles. Trunk muscles are categorized either as flexors or extensors (Boos N. et al., 2008). Abdominal muscles are the flexors, while the back muscles are the main extensors (Boos N. et al., 2008). The biomechanical role of trunk muscles is in transferring a load directly between the pelvis and the thoracic cage to the lumbar spine (Bergmark A, 1989). A group of muscles, including local and global groups, work together to stabilize the lumbar spine (Boos N. et al., 2008; Danneels LA, 2001). The local muscles such as the transverse abdominis, the deep lumbar multifidus, and the psoas, which are attached directly to the lumbar spine, are responsible for controlling the curvature of the lumbar spine as well as stiffening the vertebral segments, thus limiting motion of lumbar spine (Boos N. et al., 2008). Minor dysfunctions of those local muscle may lead to poor spinal control, indirectly inducing pain due to abnormal lumbar motion and instability. The major muscle groups that are relevant to lifting activity affect the lumbar spine, as well as muscle fascicles of the muscle groups wrapping against the bone structures (Arjmand et al., 2006; Zeem et al, 2007; McGill et al., 1986, Nussbaum et al., 1996). The rectus abdominis (RA), internal (IO) and external obliques (EO) and the lumbar erector spinae (LES) (spinalis, longissimus, and iliocostalis) muscles, which are not attached directly to the lumbar spine, but produce torque, transferring load directly between the thoracic cage and pelvis, are the global muscles (Bergmark A, 1989; Boos N. et al., 2008; Daniels LA, 2001). Contraction of the global muscles provides spinal rigidity (Boos N. et al., 2008). The global muscles are responsible for transferring force loads from the upper body to the lumbar region, relying on the local muscles to assess the lumbar stability, stabilizing the lumbar region dynamic motion (Bergmark A, 1989). Contractions are either unilateral or bilateral. Unilateral muscle contractions support bending or rotation of the vertebral column (Boos N. et al., 2008). Bilateral muscle contractions flexes extending the vertebral column (Boos N. et al., 2008). The co-contraction of antagonistic trunk muscles during lifting suggests that those muscles play an important role in supporting the stability of the spine (McGill and Cholewicki, 1996). The contraction and co-contraction of the trunk muscles both influence stability and mobility of the trunk and determine the distribution of compressive and shear forces in the lower back. (Simon et al., 1985). Repetitive, prolonged, antagonistic co-contraction of the trunk muscles and spinal muscular contractions, working to sustain the stability of the spine, may lead to chronic lower back pain (Panjabi, 1992; Cholewicki, 1993). It has been suggested that intra-abdominal pressure also contributes to the stability of the spine (Tesh et al., 1987; Dietrich M and Kedzior, 1990). This stabilizing mechanism is controlled by the neural system (McGill and Cholewicki, 1996).

2.4 Methods for estimating safe maximum loads

Researchers have developed specific assessment methods to estimate the maximum loads from specific lifting conditions, to minimize the risks of lower back injuries (Fogleman and Smith, 1995; Mital, 1984; Snook and Ciriello, 1991). The three major assessment methods may be categorized as biomechanical, physiological, and psychophysical.

2.4.1 Biomechanical Method

Biomechanical modeling is the study of the internal mechanical response of body segments in relation to their external physical work activity. Detailed anatomical models of the lumbar spine have been developed to estimate the forces and moments generated in the spine for lifting tasks (McGill and Norman, 1986; Chaffin et al., 2006; Marras et al., 1997; McGill et. al 1986). External physical elements that are related to a biomechanical analysis are the magnitude and direction of forces (or weight), specific location of the forces, the body posture during the work activity, and body movement dynamics (velocity and acceleration). Biomechanical modeling utilizes mathematical engineering principles and Newtonian mechanics. Published biomechanical models are either static or dynamic, and either two-dimensional or three-dimensional (Ayoub and Woldstad, 1999). To calculate forces from a static model, it would require knowledge of the subject's posture (segment links), the mass and length of each segment along with the location of the center of mass for each segment (Ayoub and Woldstad, 1999). For dynamic models, it would require the same data as in the static model, along with the angular joint accelerations, the linear acceleration of each segment at their center of mass, and the moment of inertia of each link through the center of mass (Ayoub and Woldstad, 1999). To estimate the failure load of the spine, the tolerance of the spinal structures has been derived from cadaveric or animal studies (Adams and Dolan, 1996). The tolerance of the spine may be estimated from functional motion segments, two vertebrae and all of the connected tissue, muscles, intervertebral disc, and ligaments. The two segments are mechanically stressed until one or many of the elements fail.

Electromyography: One important tool that is often used as part of a biomechanical assessment of physical strain is electromyography, or EMG (Chesler and Durfee, 1997; Luttmann et al., 2000; Stern et al., 2001). The utilization of surface electrodes is an electromyography (EMG) assessment modeling technique. When a group of muscles require assessment in MMH activities, surface electrodes are commonly used due to the ease of use, non-invasiveness and low expense (Lee, 2002; Mathews, 2007; Yoon, 2008; Marras et al., 1984; Deluca, 1997). Small surface electrodes taped directly on the muscle region of interest detect the twitching of muscle fibers, recording the sum of all motor unit potentials and providing a measure of spatial and temporal summation of muscle fiber activities (Marras et al., 1984; Chaffin et al., 2006). To estimate the forces generated by the contraction of specific muscles of a body part, such as the lumbar spine (L5/S1 disc), an electromyography (EMG) biomechanical assisted model may be utilized (Chaffin and Anderson, 1991; Nachemson and Elfstrom, 1970; Szabo and Chidgey, 1989; Rempel et al., 1994). There is a positive correlation between EMG activities and muscle forces acting in parallel: an increase in muscle force contraction is associated with an increase in myoelectric activity of the relevant muscle (Chaffin et al., 2006). The summation of the electric signals detected by EMGs may be used to estimate muscle forces which can help us to determine compressive and shear forces in the spine and joint moments, using biomechanical models (McGill and Norman, 1986; Chapman and Troup, 1982; Stokes et al., 1987; Chaffin et al., 2006; Marras et al., 1997; McGill et. al 1986).

An EMG-biomechanical assisted model would be an important tool in the present study to determine muscle activity during lifting of unstable loads (loads with shifting center of mass).

2.4.2 Psychophysics

Psychophysics deals with the relationship between humans' sensations and the physical stimuli (Ayoub and Woldstad, 1999). Psychophysical methods involve self-determination and personal perception of recommended weightlifting capacity. The human sensory system can function as an efficient instrument to self-evaluate the perception of workload. Psychophysical methods are the tools for measuring perception and performance (Borg 1962). Snook and Irvine (1967), researchers at the Liberty Mutual Research Center, started psychophysical experiments to determine the recommended weight limit (RWL) for lifting tasks (Snook and Irvine, 1967). They focused on lifting since it is the most common MMH task. They determined a RWL from serverAL lifting conditions. Further psychophysical research on less strenuous manual material handling activities, such as pushing, pulling, and lowering came afterwards (Snook et al., 1970; Snook and Ciriello, 1974, 1991; Snook, 1978; Mital, 1984a, 1984b; Mital and Fard, 1986; Smith et al., 1992).

2.4.3 Physiological Method

The physiological method of assessing MMH capacities of workers involves the measurement of energy expenditure (metabolic analysis) while performing a task. The method is more relevant where the cardiopulmonary system is taxed, rather than the musculoskeletal system (biomechanical). This method measures energy expenditure by the worker, indirectly from the amount of oxygen used by the body at work. Heart rate measurements while working are sometimes used for a quick estimation. The physiological assessment method may be used to determine the maximum work intensity guideline for specific tasks (Ayoub and Mital, 1989). A person's endurance in an MMH task is limited by the capacity of the body's oxygen transportation system (Ayoub and Mital, 1989). Thus, an increase in muscle activity implies an increase in metabolism, demanding higher levels of oxygen as the main source of energy. Higher demands for oxygen call for an increased in respiratory function, pumping more oxygenated blood to the muscles (Woldstad et al., 2007). The amount of oxygen used during a lifting task is linearly related to energy expenditure. Therefore, we can determine the intensity of a lifting task by measuring the amount oxygen uptake (Brown, 1971; Garg, 1976). Measuring heart rate and oxygen consumption provides an overall indication of the physiological stress placed on the body. This assessment method is complex and time consuming, which made it an unpopular method for assessing lifting tasks (Stern et al., 2001).

2.5 The NIOSH Equation

The National Institute for Occupational Safety and Health proposed the NIOSH Lifting Equation (NIOSH, 1981) in order to manage and minimize the risk of injury or lower back strain (Waters et al., 1993). The revised NIOSH lifting equation provides a practical way to estimate the potential risks of various lifting conditions in workplaces in order to reduce environmental risk factors (Waters et al., 1994). It is based on data from biomechanical, physiological, and psychophysical research. The equation calculates a recommended (manual) lifting weight limit (RWL) for an 8-hour job, under realistic lifting conditions, defined by size multipliers (independent variables for task conditions). The model specifies a constant weight as the RWL under perfect lifting conditions. The RWL is reduced, as indicated by the values of the task variables, as the lifting tasks become more stressful. The six variables that determine RWL are the horizontal and vertical locations of the load at the start of the load, vertical range of the lift, frequency of lifting, lifting duration, asymmetry, and coupling (Waters et al., 1994). The RWL is calculated from:

$$RWL\ (kg) = 23\ kg \times HM \times VM \times AM \times DM \times CM \times FM$$

where HM, VM, AM, DM, CM, and FM are respectively, horizontal, vertical, asymmetry, distance, coupling, and frequency multipliers based on their respective variable values. These multipliers are calculated with respect to the horizontal (H) and vertical (V) positions of the handled load at the start of the lift, the load transfer asymmetry to the body's mid–sagittal plane (A), the vertical distance of the lift (D), the hand-handle coupling, and the lifting frequency within the duration of the lifting activity. Distances in the above model are in centimeter. The multipliers range from 0-1, depending on lifting conditions, with 1 representing a perfect lifting condition and 0 a condition where lifting is not possible. Under perfect lifting conditions, therefore, a person should not be lifting more than 23 kg.

Even though the NIOSH lifting equation is considered the only comprehensive assessment tool to prevent work related lower back strain (Waters et. al 1993) and was designed to meet specific lifting-related criteria, there are obvious limitations within the equation. The revised equation does not account for several risk factors such as load dimension and type of load, operational process (high speed lifting-faster than 30 inches per second, lifting techniques, or knowledge of load type) or personal factors (gender, age, and physical fitness, history of injury, personality and lifting experience) (Waters et. al 1993). The revised lifting equation also does not include task factors to account for unpredictable conditions such as falls, slips, or unstable or unexpectedly heavy loads. Studies that may attempt to include any of these omitted variables would require additional biomechanical analyses to assess the physical stress on the lumbar spine (L5/S1 disc) during lifting. The NIOSH equation could possibly also be improved by including more complex lifting tasks as described above (Waters, 1994; Lee et al., 2002; Matthews, 2007; Van Dieen et al., 2001; Van Dieen et al., 2003). The present study addresses lifting a load with a shifting center of mass and may contribute to generating data and knowledge that may be relevant to a further revision of the NIOSH lifting equation.

2.6 Unexpected or Sudden Loads

Sudden forces on the body are recognized as one of the primary occupational risk factors for lower back injuries (Andersson, 1988; Marrass et al., 1993). They may be caused by unexpected loading (weights) on the body and usually results in loss of control (McCoy et al., 1997; Lavender, 1989, Manning et al., 1984). When lifting under sudden loading, there is an increase in trunk muscle contraction and co-contraction, and displacement of the trunk (McCoy et al., 1997; Lavender, 1989, Manning et al., 1984). The resulting acceleration in the trunk movement in lifting may lead to spinal tissue deformation and a higher risk of lumbar spine injury (Tsai, Lin, & Chang, 1998). Sudden loads activate passive muscles and tendons to resist the load and stabilize the back. The contraction of those passive muscles generates forces greater than necessary throwing the lumbar spine into a distress state and triggering an increase in the spinal loading rate that is positively correlated with the external loading rate (Fathallah et al., 1998). The musculoskeletal system, reacting to stabilize the lumbar spine from the effects of the sudden load, causes an increase in muscle activity and spinal compressive forces. Watanabe et al. (2011) investigated trunk muscle activity when lifting an object of greater weight than expected to determine if it would have had a higher impact on the lower back in comparison with a predetermined weight load. They found that when the subjects were aware of the weight of the object to be lifted, the activity of the external oblique, transversus abdominis, erector spinae, and lumbar multifidus muscles increased instantaneously after the start of lifting, but when the subjects were not aware of the load weight, there was a delay in muscle activity, and this may be related to the onset of lower back strain (Watanabe et al., 2011).

On many occasions, sudden loading on the spine have been caused by lack of preparedness of the lifters (McGlothlin et al., 1996) and may increase the possibility of lower back injuries. Lifters' lack of preparedness causes them to lift loads without proper assessment of the most appropriate lifting technique, producing the unexpected sudden forces that are absorbed by the back (McGlothlin et al., 1996). Moyers et al. (2003) found that, in trials without warning, the increase in muscle response was doubled when there was a sudden load change. Preparedness for an expected load weight, and the knowledge of the dimensional size of the load help lifters handle the material being moved with better care (Meyers et al., 2003). As Meyers et al. (2003) stated, an incorrect knowledge of a load may predispose a lifter to additional risks. When the characteristics of the loads are harder to be judged by the lifter, lifts are also likely to be done at a faster pace (speed lifting), resulting in higher moment forces on the lumbar spine and loss of balance (Meyers et al., 2003), and lifters who also underestimate loads are also more likely to lose their balance (Meyers et al., 2003). This kind of speed lifting, voluntarily or involuntarily, has a direct effect on spinal loading and trunk displacements, and has been shown to have a higher risk for lower back strain (Davis and Marras 2000; Davis et al. 1998; De Looze 1994; Dolan and Adam 1993; Dolan et al. 1994; Granata and Marras 1993; Granata and Marras 1995; Lavender et al. 1995). The requirement for speed the lifting process at the start causes a change of static to dynamic energy, and vice versa (McGill and Norman, 1985).

2.7 Speed Lifting

Speed lifting implies, lifting loads at fast speeds, and should not be confused with lifting frequency (Danz and Ayoub, 1992). The lifting frequency is a variable used by the NIOSH lifting equation referring to the number of times a lift is executed per minute (Danz and Ayoub, 1992). Studies that have incorporated speed as a dependent variable either use quantitative or qualitative methods. Quantitively set methods define different variable speeds by setting a timer based on the total lifting distance, utilizing a verbal and/or auditory feedback, such as metronome or voice, to start and stop the lifting task. Qualitative lifting methods rely on subject perception of what he/she considers slow, normal or fast lifting, without using a quantitative scale.

Researchers have found that speed lifting is the preferred method by experienced lifters (Valkenburg et al., 1982; Bigos et al., 1986). Workers have claimed that they felt less stress during faster lifting speeds compared to normal speeds (Garg et al.,1994), because speed lifting was found to reduce the physical work on the body (Yoon et al., 2012) and can provide enough kinetic energy during the early phases of lifting to take the load past the individual's weaker lifting levels (Ayoub and El-Bassoussi, 1978). Even though it is the preferred speed method for lifters, speed lifting has its disadvantages. Studies have shown that speed lifting exacerbates lower back stresses (Buscek et., 1987; Fathallah et al., 1998), due to the increase of inertial forces caused by an acceleration phase (Bernard and Ayoub 1999. Fatallah et al. (1998) found that speed lifting had a greater negative impact on the lumbar spine in comparison to increasing the actual weight of the load. They also found a significant relationship between lift speed and spinal loading rate during varying-speed lifting tasks:

As the speed of the lift (external load rate) increased, spinal loading rate increased. Increased speed of lifting produces greater erector spinae muscle activity, and greater compressive and shear forces (Ayoub and El-Bassoussi, 1978; Hall, 1985; Dolan and Adams, 1993). It also significantly decreases the torque-producing capability of the trunk muscles (Marras et al. 1985) and reduces the peak dynamic strength of the back (Kumar et al. 1988). Dolan and Adams (1994) showed that speed lifting influenced the peak extensor moments on the lumbar spine. Tsuang et al. (1992) examined flexion-extension and measured moments at the L5/S1 disc and hip joint, using different weight loads (50N and 150N) at different speeds (normal and fast), and found a 95% increase of the hip joint moments, when lifting the 50N at faster speeds compared to the normal speed. Speed lifting increases muscle force accelerations, thus increasing the inertial forces which directly increases compression forces on the lumbar spine (Ayoub and El-Bassoussi, 1978; Hall, 1985; Dolan and Adams, 1993; Fathallah et al., 1998; Kumar et al. 1988; Marras et al. 1985; Buscek et., 1987; Dolan and Adams, 1994; Tsuang et al., 1992).

2.8 Unstable loads with changing center of mass

An unstable load refers to one in which the center of mass changes with movement of the load as it is lifted, lowered or carried. Both liquid and solid loads may exhibit this instability. The shifting of the center of mass in a solid may be due to parts of the load mass moving relative to the rest of the mass (e.g. the contents of a trash bag moving around in the bag while being carried), while the shifting in a liquid may be due to the liquid mass moving around within a container (e.g. a water in a half-full container sloshing around as the container is moved). The center of mass of an object being lifted by a person may be displaced not only in the frontal plane of the person, but also in the sagittal plane, creating awkward and complex three-dimensional moments on the lumbar spine (Meyers et al., 2003).

Manual handling of unstable loads has become very common in many industries, mostly in the foods and beverage industry, and has resulted in a higher incidence of low back injuries compared to the handling of stable loads with similar lifting demands (McGlothlin, 1996). Similar industries such as convenience stores, food deliveries, or construction that involve lifting unstable loads have also shown a higher incidence of low back injuries (Marras et al., 2009; Vivek et al. 2000; Bernard, 1997). Several researchers have attempted to explain these injury effects in terms of changes in muscle activities and forces within the spine (Lee and Lee, 2002; Matthews et al., 2007; van Dieën et al., 2003; McGill et al., 2004).

A solid shifting load within a container (e.g. freely moving solid ball within a box) will have a center of mass that might shift suddenly and impact the internal walls of the container (Lee and Lee 2002). A shifting liquid load has a different random moving pattern and a gradual shifting center of mass (Van Dieen et al., 2001; Pinto, 2001), and different types of liquid viscosity levels would trigger different speed rates of the shifting center of mass, and movement patterns.

Unstable liquid loads within a container, when lifted, may generate internal forces that pose a high risk of injuries to lifters' lower backs (L5/S1 disc) and are more damaging, in general, to the musculoskeletal system (Bakker et al., 2007 and Norman et al., 1998; Lee et al., 2002). Liquid fluidity permits the continuous unpredictable internal shifting behavior of the load's center of mass, challenging the lumbar spine multiple times (McGill and Norman, 1985). The exacerbated effects are similar to those sustained when lifting a load with an unexpected weight. The shifting center of mass implies that the characteristics of the load are difficult to judge for stabilization (Cholewicki et al., 1997).

Sudden loads may have a single unexpected negative force impact on the lumbar spine (Marras, 1998), and unstable loads may have multiple sudden impacts. The forces generated by the shifting center of mass throws experienced lifters off guard, stressing their lumbar spine and putting their system in prolonged distress (Cholewicki, 1993; Marras et al. 2013; Watanabe et al., 2011). The lower lumbar spine may be strained due to the continuous unexpected load disturbance created by the continuous shifting of center of mass in unknown directions (van Dieën et al., 2003). The continuity of lumbar spine disturbance occurs because of the erratic reactionary response of the muscles trying to stabilize the body, responding to the shifting load, not knowing whether to flex or contract. Every time the lifter responds toward one side of the load, the center of mass may shift to the other side. Multiple sudden and alarming events associated with unstable load shifting cause a reflex overreaction of the back-extensor muscles, which substantially increases spinal compression loading (Cholewicki, 1993; Marras et al. 2013). That sudden alarming event continues to occur until the unstable load settles.

As noted earlier, sudden loads could cause large forces on the lumbar spine and tissues to inflame, worsening back fatigue and stress on workers (Lee et al., 2002). Worse effects may occur when dealing with unstable loads. Lee et al. (2002) explored the physical risk associated with handling an unstable load, using a biomechanical measurements model. They concluded that the nervous system reacts to an unstable load by stabilizing the joints closer to the direction of load-shift. Their experiment has shown that the central nervous system responds within 0.2 seconds with maximal contractions of 57.2% of the erector spinae, after the impact, to stabilize the joints immediately. Lee et al. (2002) also suggested that future research should be extended to include unstable loads with more complicated tasks triggering a wider range of muscular activities. In the present study our extension includes asymmetric lifting at different speeds.

2.9 Unstable Loads and Muscular Actions

Sudden changes in a load activate both reflexive and voluntary muscles. Initially, trunk muscles contract to increase stability. The contraction increases spinal compression especially when the load changes (Granata and Marras, 2000; Marras et al., 1987). Van Dieën et al. (2003) monitored the internal/external oblique and rectus abdominis muscles to determine the co-contraction of the antagonistic muscles of the back. They noticed that the lumbar spine contraction and compression force levels worsen during the lifting of liquid loads (unstable loads) and the antagonistic muscles co-contract to support the back. They used the lifting of unstable loads to further test the proposition that abdominal co-contraction serves to increase trunk stiffness during lifting (Granata and Marras, 2000). Loads with a moving center of mass (sloshing water) were used to create a situation in which an irregular movement of the trunk during lifting can occur. Those movements cause unpredictable force moments about the intervertebral joints and threatened spinal stability (Cholewicki and McGill, 1996). The antagonistic co-activation contracts to counteract the threat and would be higher when lifting the unstable load as compared to a stable load (Cholewicki et al., 1995). Meyers et al. (2003) tested the lumbar spine muscle activity, using electromyography (EMG), by changing the stable load within a bin at a predefined location, and altering the load's center of mass. They found that there was a signal peak in the upper erector spinae muscles of the lower back and the oblique's when the weight was placed on the contralateral sides of the bin, closer to the body. Myers et al. (2003) found muscular spike contractions of upper and lower erector spinae when the center of mass of the load was closer to the body (Meyers et al., 2003), and the different center of masses per load had different effects on the lumbar spine. These results were contrary to conventional wisdom of lifting loads closer to the body to lessen lower back strain but was probably due to lifters lifting the load faster when it was closer to the body. Lifting loads at a faster pace shortens the muscles' contraction velocity causing greater EMG responses (Meyers et al., 2003).

Most unstable load studies have focused on the load type effects (stable vs. unstable) on the lumbar spine (Mathews, 2007). Most researchers agree that lifting unstable loads potentially strains the lumbar spine, leading to low back injuries (Van Dieen et a., 2001,2003; Lee and Lee, 2002; Mathews et al., 2007; Pinto et al., 2013). These studies were limited to the focus of load type in a symmetrical lifting style (Van Dieen et a., 2001, 2003; Lee and Lee, 2002; Mathews et al., 2007), but none has been comprehensive enough to cover other risk factors associated with the load type (Pinto et al., 2013), such as the effects of asymmetric lifting of the unstable load, speed lifting, or other variables that influence the shifting of the load. Only recently, Pinto et al., (2013) researched the effects of using control designs when lifting unstable loads in an asymmetrical style. More research is needed with unstable loads for investigating the variety of variables that are known to affect the musculoskeletal system when handling stable loads (Burg et al. 2001). In addition, investigations should not be limited to symmetric lifting. Asymmetric lifting of unstable loads at different speeds deserves urgent attention. This is the focus of the present study. There is also a need to compare the effects of load type (stable vs. unstable loads), simulating real life scenarios.

3. Mechanics of asymmetric lifting

Prior research on unstable loads focused on symmetrical motion of the body, even though unstable load handling involves significant asymmetrical movements (Lee et al., 2002; Van Dieen et al., 2001; Van Dieen et al., 2003). Lifting a load asymmetrically exacerbates the negative effects of spinal compression and shear forces. To further understand the effects of asymmetrical effects of lifting unstable loads, it is critical to understand the mechanics of basic asymmetric lifting of stable loads and its effects on the lumbar spine. The present study aims to examine the effects of asymmetrical unstable load lifting.

During asymmetric lifting of stable loads, axial trunk rotation and lateral bending have been shown to increase the risk of LBP, with concomitant increases in torsional and spinal shear loadings (Waters et al., 1993; Meyers, 2003), making the body more susceptible to lower back problems. Past studies on unstable loading in symmetrical and asymmetrical ways have shown that simply twisting is considered damaging to the spine (Adams and Hutton, 1981), because during flexion-rotation the anterior annulus fibrosus is stretched , causing it to become thinner, pressuring the nucleus pulposus towards the weaker sheath of posterior elements (Bogduk, 1991). Andersson et al. (1977), Schultz et al. (1979) and Ortengren et al. (1981) stated that intra-discal pressure and the intra-abdominal pressure both increased when the trunk was loaded in rotation. Twisting of the trunk during lifting may be the preferred method by most lifters and requires less energy expenditure than moving the feet without twisting the trunk, but it leads to a much higher risk to lower back injuries (Simon et al., 1985). Twisting the trunk during lifting of a load increases the intra-discal and intra-abdominal pressures when the trunk is loaded in rotation, and may damage the spinal structures (Anderson et al., 1977; Schultz et al., 1979).

3.1 Simulating Load Instability in a Laboratory

Table 1 below presents a summary of four publications related directly to lifting unstable loads. The five studies' objectives were to determine the negative impact of load type (stable vs. unstable) on the back. All five studies used only symmetrical lifting tasks along the sagittal plane (Van Dieen et al. 2001 and 2003; Lee and Lee, 2002; Matthews et al., 2007). Van Dieen et al. (2001 and 2003) determined the negative impact on the back by estimating compression and shear forces on L5/S1 disc using EMG electrodes on the bilateral external oblique and lumbar erector spinae. The other studies used Normalized EMG (NEMG) to determine any change of specific muscle contractions (bilateral external obliques, lumbar erector spinae, and latissimus dorsi), related to supporting the back (Lee and Lee, 2002; Matthews et al., 2007).

Table 1: Past Research Papers Summary on Unstable vs Stable Loads

Reference	van Dieen et al., 2001	Lee and Lee, 2002	van Dieen et al., 2003	Matthews et al., 2007
Research Objective	Load Type	Load Type	Load Type	Load Type, External Weight
Load Type	Unstable Load – Water Stable Load - Weight taped at bottom container	Unstable Load - Pin on ramp inside container Stable Load - fixed weight at	Unstable Load – water Stable Load - Ice	Unstable Load - Weight Hung from chain link Stable Load - Fixed Weights
Lifting Style	symmetric	symmetric	Symmetric	symmetric
Total Lifting Weight	22 lb	40 lb	33 lb	22 lb
Lifting Speed	Synchronous On the Beep	Lifter Preference	as fast as possible	six lift per minute
Method Used	EMG - bilateral External Oblique, Lumbar Erector Spinae	EMG, Unilateral Lumbar Erector Spinae	EMG - bilateral External Oblique, Lumbar Erector Spinae	EMG - bilateral External Oblique, Lumbar Erector Spinae, Latissimus Dorsi
Result Values	Estimated Compression and shear forces on L5/S1 disc	Peak NEMG	Estimated Compression on L5/S1 disc, Mean abdominal Co-activation	Peak NEMG, Mean NEMG
Overall Average Impact Affected by Unstable Load	~5000N	~63% MVC	2000N - 8000N	~45%
Change Increase In Lifting Unstable Load Compared to. Stable Load	50th Percentile = 255 N 95% Percentile = 458 N 50th Percentile = 5% 95% Percentile = 9%	20%	Phase 1 = 16N	External Oblique = 22 pounds = 5%
			Phase 2 = 667N	Lumbar Erector Spinae= 33 Pounds = 9%
			Phase 3 = 266N	Latissiumus Dorsi= both Pounds = 0.9%

Not all these studies used water to simulate unstable loads. Liquid water was used as the unstable load type for three of the four publications (Van Dieen et., 2001,2003; Lee and Lee, 2002). Mathew et al. (2007) utilized a weight hung from a chain link, while Lee and Lee (2002) used a rolling pin inside the container.

In different studies, Van Dieen et al. (2001; 2003) tested lifting liquid water versus solid weights. The core difference being that the first study was with controlled timing using a metronome (Van Dieen et al., 2001), while the second requested the subjects to lift as fast as possible. Both studies showed an increase in the compression force of the L5/S1 disc of 9% and 15% respectively, using EMG estimations.

Lee and Lee (2002) used a 12kg iron pin roll locked in place at the top of a container. The pin roll unlocks to roll down the ramp at a 12-degree angle against the container wall when the container was lifted. The purpose of their simulation was to have a noticeable sudden impact on the lumbar spine. The result showed that the impact increased the lower back muscle activity to approximately 22% of the maximum voluntary contractions compared to a stable load. The equipment had a rolling pin that would always roll in the same direction, but which eliminated the randomness of the shifting of center of mass expected from unstable loads. The arrangement was only effective for symmetric lifting, impacting the same wall in the container. This implies the center of mass would always shift inwards towards the lifter.

In a study of risk management in maritime industries, Mathews et al. (2007) used a weight attached to a bar by a chain. The weight could move freely, simulating the effects of a moving platform and load instability for a common lifting task

Meyers et al., (2003) failed to test the effect of the shifting center of mass shifting during lifting. The center of mass was pre-set prior to the subject lifting the load (Figure 3). Lifters' lack of knowledge of where the center of mass was located did not alter the average or peak EMG. This could have been due to lifters sensing heavier loads at either side of the bin during the initial lifting phase, allowing them to adjust instantly.

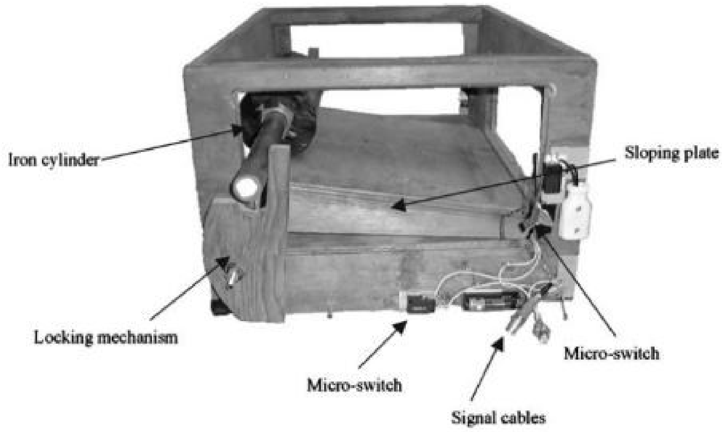

Figure 1: Lee & Lee (2000) -Unstable Load Rolling Iron Cylinder Bin Model

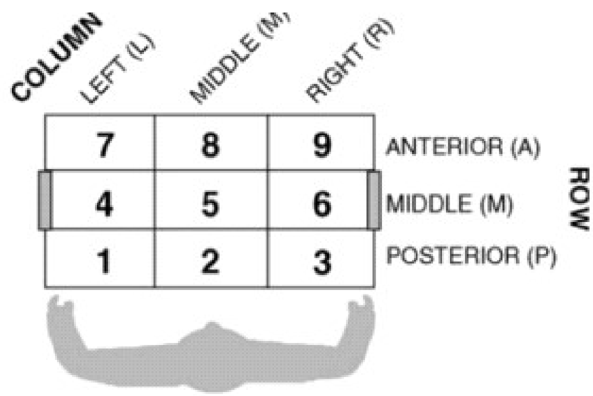

Figure 2: Meyers et al., (2003) - Lifting Bin Model

3.2 Common Methods Utilized for Unstable Loads

Many of the previously mentioned studies used surface electromyography, EMG, to measure muscle activity responses during MMH activities. With technology advancement, 3D image rendering of subjects doing MMH tasks with supporting of EMG measurements provide a more accurate estimates of muscle responses and L5/S1 disc compression and shear forces. All the tools except “Anybody Modeling System” were easy to use (Table 2). While they all claim to work for asymmetrical lifting styles, they all had different limitation, such as flexion and the number of handles on the bin being lifted. Summarized below are the results of the different biomechanical imaging tools that have been utilized in research (Table 2).

Rajaee et al., (2014) published a comparative evaluation of several quantitative biomechanical lifting models to estimate spine loads during static activities. The models were: the hand-calculation back compressive force (HCBCF), Simple Polynomial, the University of Michigan’s 3D static strength prediction program (3DSSPP), anybody modeling system, and Linked-segment biomechanical model (LSBM). The models have their pros and cons, and each predicted considerably different compression and) shear forces from the others at the L5/S1 and L4/L5 discs. This strongly suggests that great care must be used in selecting a model for estimating spinal forces.

Table 2: Assessment models' Effectiveness on L5/S1 disc force estimation

	The Link Segment Biomechanical Model (LSBM)	**The simple polynomial of low back compression**	**The Anybody Modelling System similar to Visual 3d**	**Regression Models**
Asymmetric Tasks	✔	✔	✔	✔
Sensitivity to load height when standing in fixed position, causing a shift in center of mass of load.	X	X	✔	X
Ease of Use	✔	✔	X	✔
Recommended application domains	Symmetric tasks except those with load held on sides in both hands	Symmetric/assymmetric tasks except very light or heavy ones	Symmetric/assymmetric tasks with small to moderate flexion	Symmetric tasks except those with load held on sides in both hands + tasks with asymmetric loading

It seems that the most effective way to determine compression and shear forces on the L5/S1 disc lumbar disc while lifting asymmetrically would be by using the visual 3D modeling with EMG analysis for evaluating muscle response activity at the L5/S1 disc. In the present study, load cells attached to a container being lifted was used help us determine any shift of the center of mass of the (liquid) load being lifted, and its effect on the L5/S1 disc.

Chapter 4. Methods

The biomechanical effects of the lifting were determined using advanced video technology, available in the Kinesiology lab, UTA, to estimate moments and forces generated in the lumbar spine during load lifting.

A biomechanical model of load lifting, which is commonly used to estimate the spinal compression, shear and torque forces on the L5/S1 disc, moments about various joints, and muscular contraction levels were utilized for the analyses as used by other researchers (Marras et. al, 1998; de Looze. et al, 1992; Chen, 2000; Gillette, 1999; Schipplein et al., 1990; Sparto et al. 1997; Trafimow et al., 1993). The biomechanical variable estimations were based on measurements of the external forces acting on the body, together with the relevant anthropometric and kinematic data. The data were combined for the link segments using equations of motion to estimate joint reaction moments and forces (Marras et. al, 1998; de Looze. et al, 1992).

According to the National Institute of Occupational Safety and Health (NIOSH, 1981) any compression forces on L5/S1 disc predicted to be higher than 3400N are considered hazardous to some workers and above 6400N to be dangerous to most workers. In this study, the NIOSH limits are used as a standard of comparison against the L5/S1 disc forces obtained in this study.

For the safety of the experimental subjects, the weight that the subjects lifted was set below that of NIOSH's maximal recommended weight limit (RWL) of 51 pounds, under ideal lifting conditions, in a typical 8-hour job. Thirty pounds (30 lb) was used for the experimental tasks. It has been shown to have no complications (Marras et. al, 1998).

4.1 Participants

Three healthy male participants (age: 23±10 years; body mass: 175lb ± 25lb) were recruited for the study. Participants had no prior history of any acute or chronic lower back pain, injuries or any relevant surgical history to legs, neck, arms, shoulders knees etc. To minimize the differences between participants, Body Mass Index (BMI) only within the range of 18.5 – 24.9 (BMI, kg/m2) were accepted.

4.2 Apparatus

Sixteen MX T40S cameras (4MP resolution 2336 x 1728) connected to Vicon T-Series motion capture system analysis (Vicon Motion Systems Ltd., Denver, CO) were used to collect data during the lifting tasks at a sampling rate of 200Hz. Two ground reaction force plates with a sampling rate of 1,000 Hz using AMTI Optima OPT400600-2000 (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used, capturing ground reaction forces (GRF) and moments synchronized with the video motion capture system. A plastic bin of dimensions (28.5" x 19.6"x15.27"), instrumented with two three-dimensional cell load force transducer handles (PCB model 261A01; PCB Piezotronics, Inc, Depew, NY) was partially filled with water to provide an unstable load (Figure 3), and the load weight was taken as the combined weight of bin plus water. Lifting reaction forces (LRF) and moments were derived from data sampled at the rate of 1000 Hz from the load cells. The PCB force transducers have a maximum tension or compression range (Z) of 4500 N; shear range X, Y of 2200 N; Z axis sensitivity 0.56 mV/N; and X and Y axis sensitivity of 2.2 mV/N. In addition, an EMG system with a sampling rate of 2000 Hertz, relayed to the Vicon system was used to synchronize the data from muscle contractions alongside with GRF, LRF and body movements. Bilateral electrical activity of the rectus abdominis, external oblique, internal oblique, latissimus dorsi, upper erector spinae, and lower erector spinae muscles (McGill & Norman, 1986) were recorded using a 16 channel Bagnoli desktop EMG system (Delsys Inc., Boston, MA).

Figure 3: Customized Lifting Bin instrumented with three-dimensional force transducers PCB model 261A01 (PCB Piezotronics, Inc, Depew, NY) which were also sampled at 1000 Hz.

Prior to electrode placement, the skin was lightly abraded and cleaned with rubbing alcohol to reducing signal impedance. Pre-amplified bipolar surface electrodes (DE-2.1 Single Differential, Delsys Inc.) composed of two silver bars (1 mm wide x 10 mm long) and a fixed inter-electrode distance of 10 mm were placed over the middle of the muscle surface located between the motor point (muscle mid-point) and the tendon–muscle interface (Figure 4,5 and 6). The longitudinal axis of the electrodes was aligned parallel to the length of the muscle fibers. A large 2-inch self-adhering Dermatrode electrode (American Imex, CA, USA) ground electrode was positioned on the medial surface of the anterior tibia. EMG signals were amplified 1000 x and bandpassed filtered (20-450 Hz) using a Bagnoli biological amplifier which has a common mode rejection ratio of 85 dB and system noise < 1.2 μV (RMS). The EMG data was sampled at 1000 Hz with a voltage range of ±5 V.

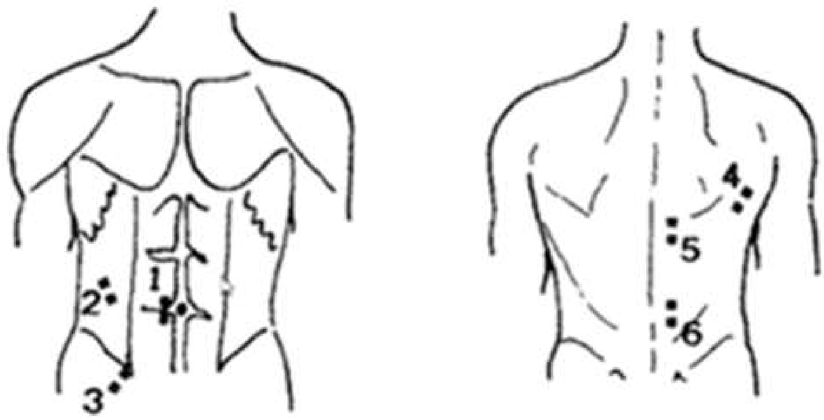

Figure 4 EMG Electrode Location on Abdomen & Back McGill SM, Norman RW. 1986 Volvo award in biomechanics: Partitioning of the l4 - l5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine*. 11: 666-678, 1986.

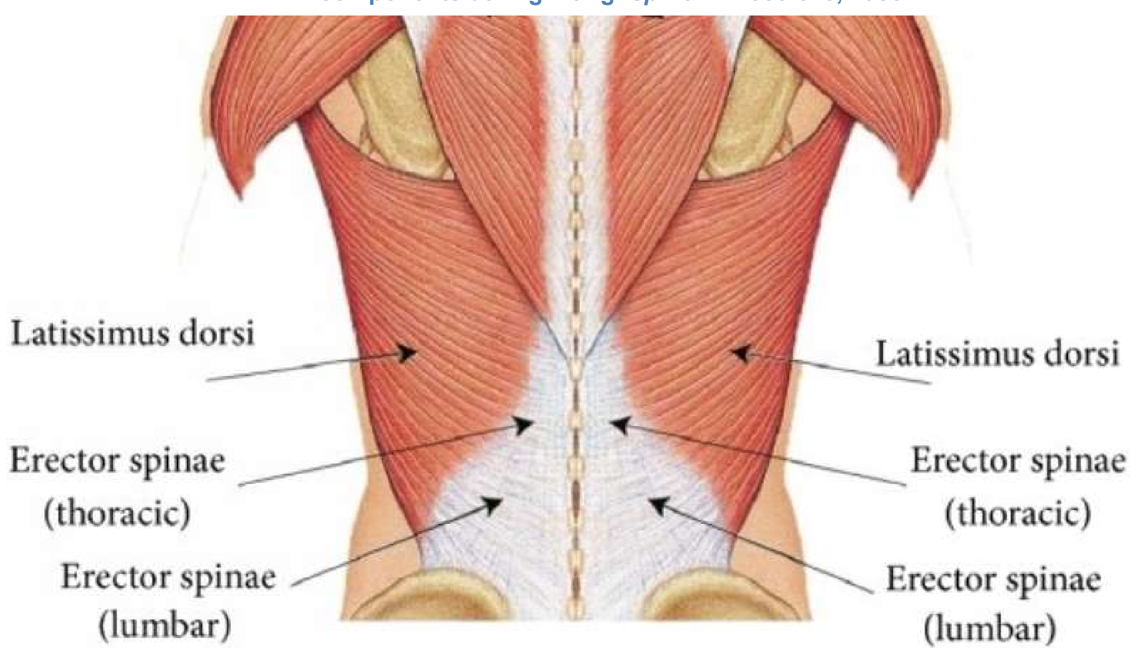

Figure 5: Back Muscles Monitored Study

(Source: https://bamboocorefitness.com/the-transverse-abdominis-the-spanx-of-your-abdominal-muscles/)

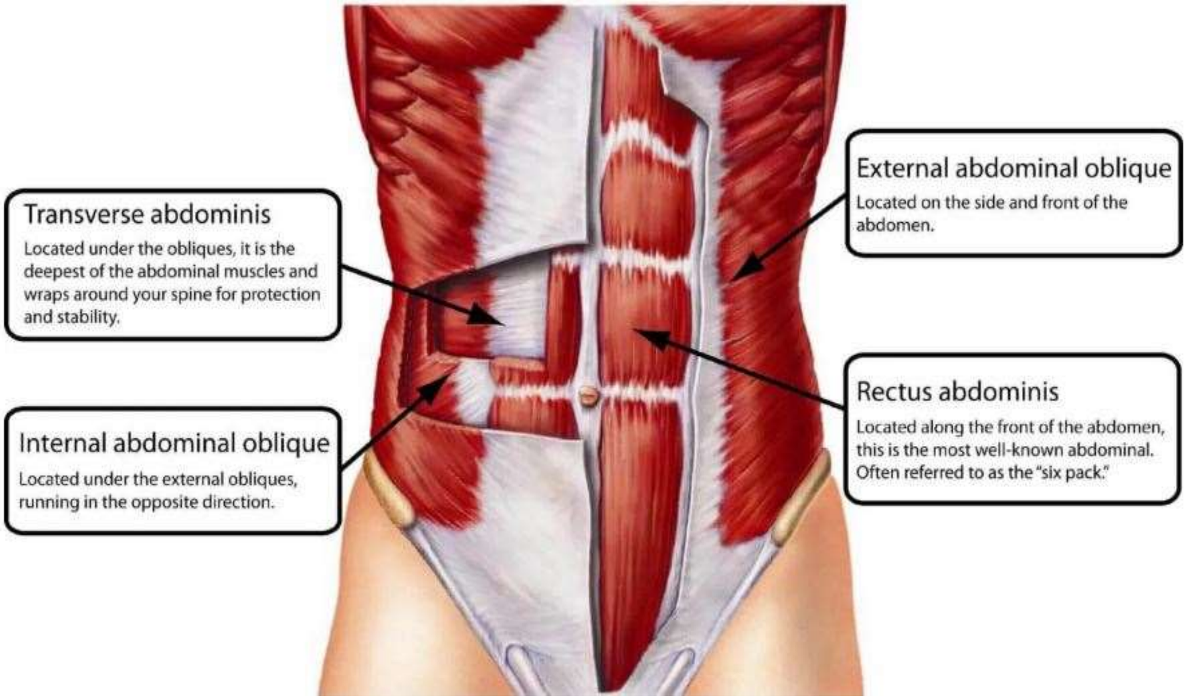

Figure 6: Abdomen Muscles Monitored for Study
(Source: https://bamboocorefitness.com/the-transverse-abdominis-the-spanx-of-your-abdominal-muscles/)

The 30lb weight for lifting was combined weight of the plastic box plus water plus attached load cells. The load cells were attached to specially designed handles that determined XYZ forces of the shifting weight load during the lift. To achieve adequate sloshing around of the water (center of mass movement) during the lifts, the box was half filled with water.

On every participant, 97 reflective markers (14mm in diameter) were attached bilaterally to the subject's skin over anatomical landmarks using simple double-sided tape. They were attached to detect body movements during the subjects' lifting of the load (Figures 6, 7, 8, and 9). Forty-five of the markers, according to Tyler's full body marker set (Table 6, Appendix A) were placed on the upper body, and fifty-two on the lower body. On the lower body the superior most point of iliac crest in the sagittal plane (RPP, LPP), anterior superior iliac spine (RAS, LAS), posterior superior iliac spine (RPS, LPS), greater trochanters (RHP, LHP), medial and lateral epicondyles of the femur (RMK, RLK, LMK, LLK), right and left thigh clusters , each cluster with 4 reflective markers, right thigh (RTH1, RTH2, RTH3, RTH4) , left thigh (LTH1, LTH2, LTH3, LTH4),left and right tibia tuberosity (RTT, LTT), right shank cluster (RSK1,RSK2,RSK3,RSK4) , left shank cluster (LSK1,LSK2,LSK3,LSK4) , medial and lateral malleoli (RMA, RLA, LMA, LLA), right and left first metatarsal (R1MH, L1MH), right and left fifth head metatarsals (R5MH, L5MH) and base metatarsals (R5MB, L5MB), right and left heels (RHL, LHL), and lastly on the right and left toe (RTOE, LTOE). On the upper body non-collinear reflective markers on molded thermo-plastic shells were placed on the posterior thorax, right upper arms (RADL, RPDL, RUA1, RUA2, RUA3,RUA4, RLEL), left upper arm (LADL,LPDL,LUA1,LUA2,LUA3,LUA4,LLEL), right forearms (RMEL, RFA1, RFA2, RFA3,RFA4, RWRR, RWRU), left forearm (MEL,LFA1,LFA2,LFA3,LFA4,LWRR,LWRU), right hand (RHR,RHM,RHU), left hand (LHR,LHM,LHU), head (THEAD, PHEAD, C7, AHEAD, RHEAD, LHEAD, RNECK, LNECK), trunk (RAC,LAC,CLAV,STERN,T2,T8RTL,LLT,RUT,LUT), and lastly on the lumbar spine (L1,L3,L5) (Figure 6, Figure 7, Figure 8, Figure 9). A standing trial was recorded.

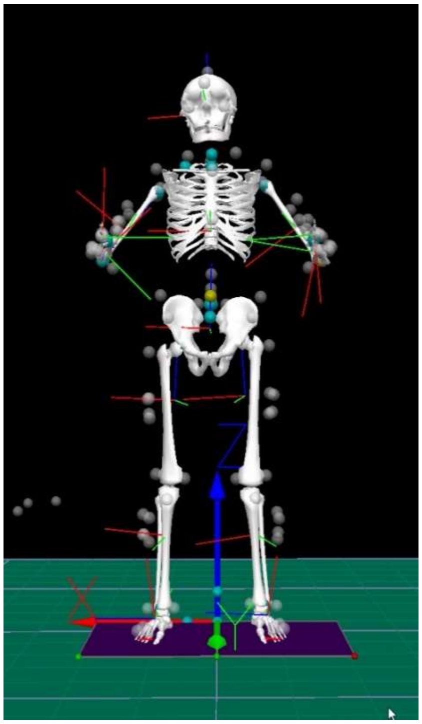

Figure 6: Full Body Front View 3D Reflective Markers Model

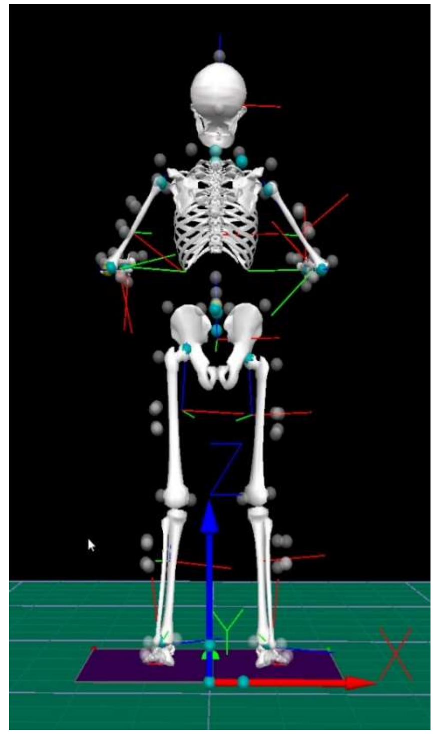

Figure 7: Full Body Back 3D Reflective Markers Model

Figure 8: Full Body Front Reflective Markers and EMG Placements On Subjects

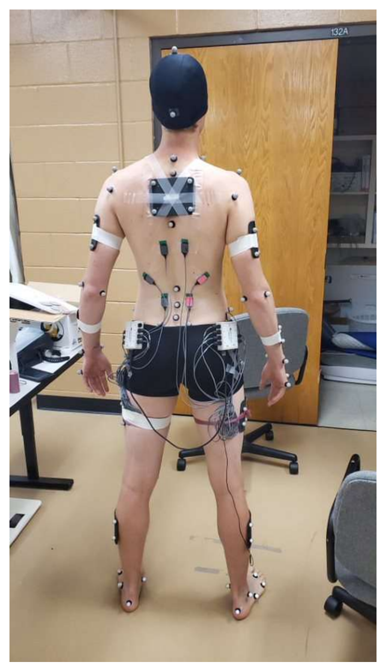

Figure 9: Full Body Back Reflective Markers & EMG Electrode Placements on Subjects

4.3 Experimental Process

Subjects were required to lift the 30lb load bin, with both hands, in eight different lifting conditions (2 load stability modes x 2 speeds of lifting x 2 modes of symmetry), while standing on solid fixed force plates in the floor. The stability modes were (i) stable load with fixed center of mass, and (ii) unstable load with shifting center of mass; the speeds were (i) normal comfortable lifting speed, and (ii) as fast as possible speed (Ayoub et al., 1999; Lavendar et al., 2003); and the symmetry modes were (i) symmetrical lifting in the sagittal plane and (ii) asymmetrical lifting in no fixed plane. The force plates measured XYZ forces generated from the subject's feet while lifting the bin. The bin was positioned on the ground in the mid-sagittal plane a few inches in front of the subject. No specific lifting style was required. Subjects were instructed to follow free-style lifting (Lavender et. al, 2003) provided it was effective and safe. This eliminated the constraints imposed by the so-called 'stoop' or 'squat lift' that some researchers have used in the past (Marras et. al, 1998).

For the symmetrical lift, the lifting trajectory was from the ground, directly in front of the subject, upward onto a table at waist level (Figure 10). For the assymmetric lift, the trajectory was from the ground upward and twisting at the waist 90 degrees, onto the table in the mid- sagittal plane (or in the coronal plane) (Figure 11). Immediately after the lift, subjects returned the bin back to ground level in the exact reverse order. For data collection, each subject repeated every lifting task at least three times. A rest break of at least 1 minute was provided between lifts for each participant (Marras et. al, 1998).

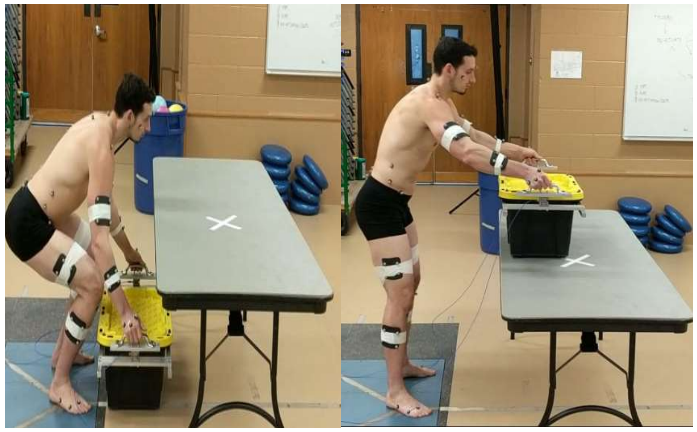

Figure 10: Lifting Task along sagittal Plane

Figure 11 : Asymmetrical Lifting Task picking up from the front along sagittal plane ground level releasing 90-degree angle from pickup

To accommodate experimental subjects of different statures, an adjustable height table was used for the lifting task. With the box on the table the handles were set approximately 56.6 ± 10 inches from the floor, based on published anthropometric dimensions of male American population.

4.4 Kinematic and Kinetic Data Analysis

Vicon software was used to simultaneously collect motion data, EMG Signals, force plate and load cell data. The data was then transferred to Visual 3D software format to estimate compression and shear forces on the lumbar spine. Three-dimensional software (C-Motion, Germantown, MD, USA) was used to process three-dimensional kinematic and kinetic data for each participant. During lifting tasks, the subject's reflective markers were not always visible or unnecessary reflective noise were captured by the cameras. Hence a software function to automate noise filtering or gap filling was utilized. Marker trajectories were used to filter with a fourth order recursive Butterworth low-pass filter with cutoff frequency of 6 Hz. Ground reaction force data was filtered using a 20th order critically damped Butterworth low-pass filter with cutoff frequency of 30. Three-dimensional joint angles were calculated using an x (flexion/extension), y (abduction/adduction), z (axial rotation) Cardan rotation sequence (Cole et al. 1993). A fifteen-link segment body model was used to estimate the internal forces (Schipplein et al. 1990). The link segments used were the foot, leg, thigh, pelvis, thorax, neck, upper arms, forearms and hands. Reaction forces and moments at each joint were calculated using inverse dynamics for the standard link segment model. Dynamic linked segment models were used to describe the forces and moments acting on the lumbar spine while lifting in the laboratory and in the industry (Chaffin and Park, 1973; De Looze et al., 1993; Freivalds et al., 1984; Lavender et al., 1999; Potvin et al., 1992; Schipplein et al, 1990). Body segment parameters (mass, center of mass location, and moment of inertia) were obtained using de Leva (1996). Hip joint center locations were obtained using Bennett (2016).

4.5 L5/S1 Disc Position and Orientation

The location of the L5/S1 disc was estimated based upon the typical size of an adult male lumbar vertebrae (Nissan & Gilad, 1986; Zhou et al., 2000). A virtual marker was created in Visual 3D to identify the location of the L5/S1 disc. A technical coordinate system was created to define the orientation of the L5/S1 disc as follows: a unit vector defining the X axis was created by subtracting the LPS marker position from the RPS marker, a unit vector defining the Z axis was created by subtracting LV5 marker position from LV3 position, the Y axis unit vector was defined the cross product of Z unit vector crossed into the X unit vector. The Visual 3D waist force was then transformed into the L5/S1 disc coordinate system (Figure 12).

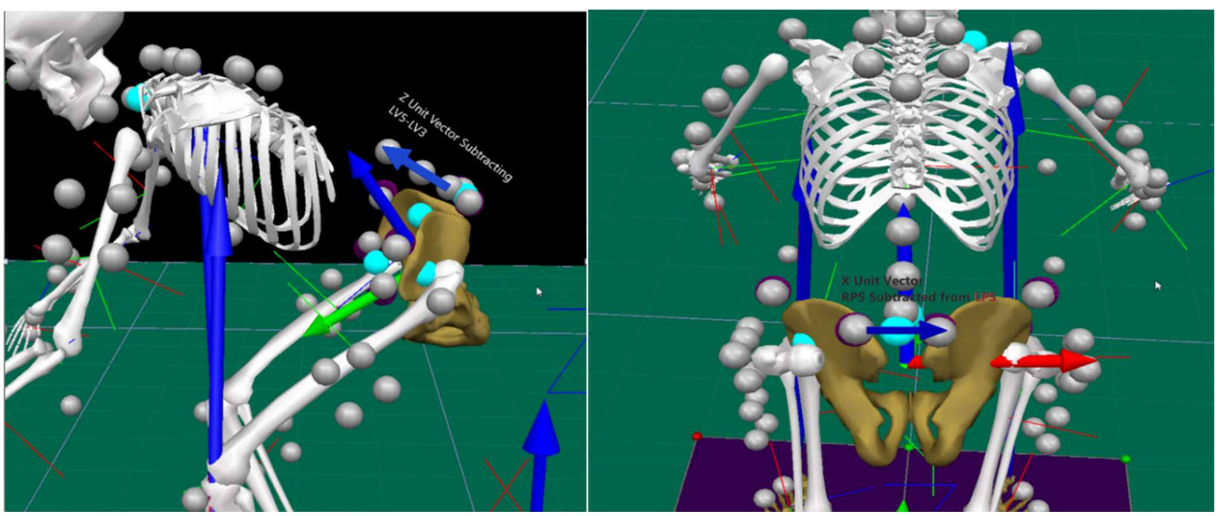

Figure 12: Technical Coordinate System Defining L5/S1 Disc Orientation

Since every subject has different anthropometric dimensions from other subjects, regression equations from de Leva were used to determine body segment parameters for men (de Leva, 1996). A landmark (virtual target) was estimated between L5/S1 vertebras (disc). Physical reflective markers are the ones taped on the subject's body and visible to the cameras. Non-visible Virtual targets called landmarks can also be created within a subject's template. Landmarks can be defined relative to a segment, along a line, on a plane, or as a projection onto a line or a plane. The landmark was used to estimate forces and moments because of net forces from force plates and the loads on the bin. A linked segment model using bottom-up approach was used to calculate joint moment at L5/SI disc during lifting (de Looze et al. 1992). The reactive forces and moments were calculated with a starting point from feet (ground reaction forces) to the hand (load cells).

4.6 Estimating Compression Force at L5/S1 Disc

The erector spinae muscles generate the extensor force for lifting as seen in figure 13. The forces generated by the erector spinae compresses the vertebras. That force acting at the superior surface of the pelvis (F pelvis) was transformed into the L5/S1 coordinate system defined in figure 12. The upper body extension moment was divided by its lever arm, r, to determine the component of lumbar compression due to the action of muscles across the abdominal joint (figure 13).

The lumbar compression force was computed as the sum of the transformed F Pelvis and F Extensors acting in the L5/S1 coordinate system.

4.8 Estimating Shear Forces at L5/S1 Disc

The shear forces at L5/S1 disc were estimated using visual 3d c-motion software. The anterior-posterior and the medial-lateral shear forces acting on the pelvis were transformed into the lumbar coordinate system at L5/S1 estimated disc location. The anterior-posterior shear force acting on the L5/S1 disc system was the Y component of the lumbar force. The medial-lateral force acting on the L5/S1 disc system was in the X component of the lumbar force.

4.9 Estimating Torque moments at L5/S1 Disc

Torque moments were estimated by transforming the waist moments into the lumbar coordinate system at L5/S1 disc. Torque moments were estimated using visual 3d c-motion software. The torsion was based off the upper body force and the angular velocity with respect to the.

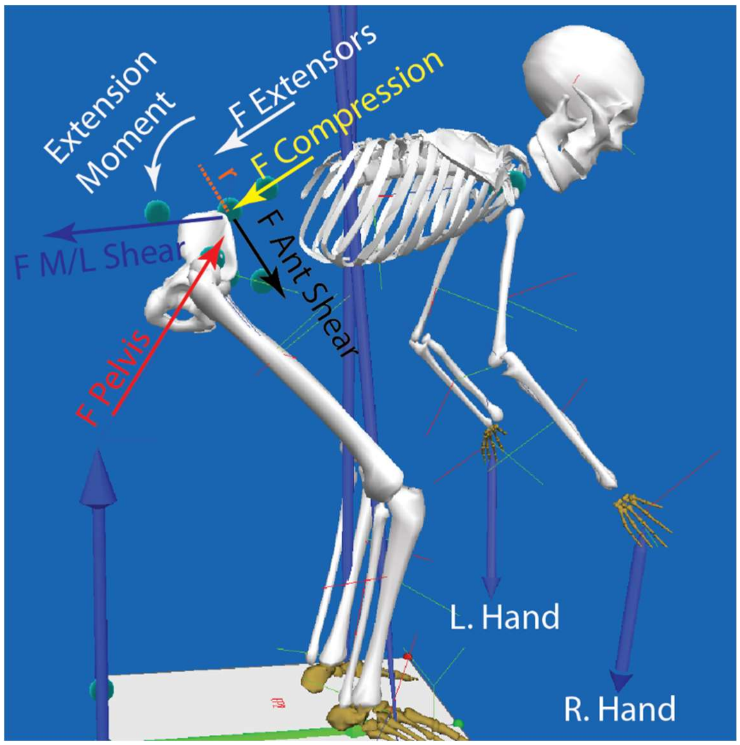

Figure 13:Lumbar Compression force (F Compression), medial-lateral force (F M/L) and Anterior-Posterior (F Ant Shear) Shear forces. Load cell forces (L.Hand and R.Hand)

Our method provided answers to the following experimental questions:

What were the estimated forces (compression, shear, and torque) on the L5/S1 disc when lifting unstable compared to stable loads? What were the estimated forces when speed lifting compared to slow lifting of a load? What were the estimated forces when lifting the same load in symmetry vs asymmetry? As stated earlier, past research has not provided a comprehensive method to sum all the forces impacting the body while lifting unstable loads.

5. Results

Peak compression forces at the L5/S1 disc were estimated following the above methodology and using Visual 3d c-motion software. All mention of compressive or shear forces and torsion moments below refer to those at the L5/S1 disc only. The peak compressive forces for one subject under eight different lifting conditions and three trials per condition are graphed below (Figure 14). As seen, the lifting speed had the highest force impact on the lumbar spine, followed by the lifting style (Symmetrical or Asymmetrical). The fast lifting speed had an average increase in the peak compression force of 579N (range=3143N to 372N) from a normal speed lifting force (data from Figures 17 and 18 combined). Asymmetrical compressive force increased by 282N, on average, compared to the symmetrical force, from 3291N to 3573N (Figures 15 and 16).

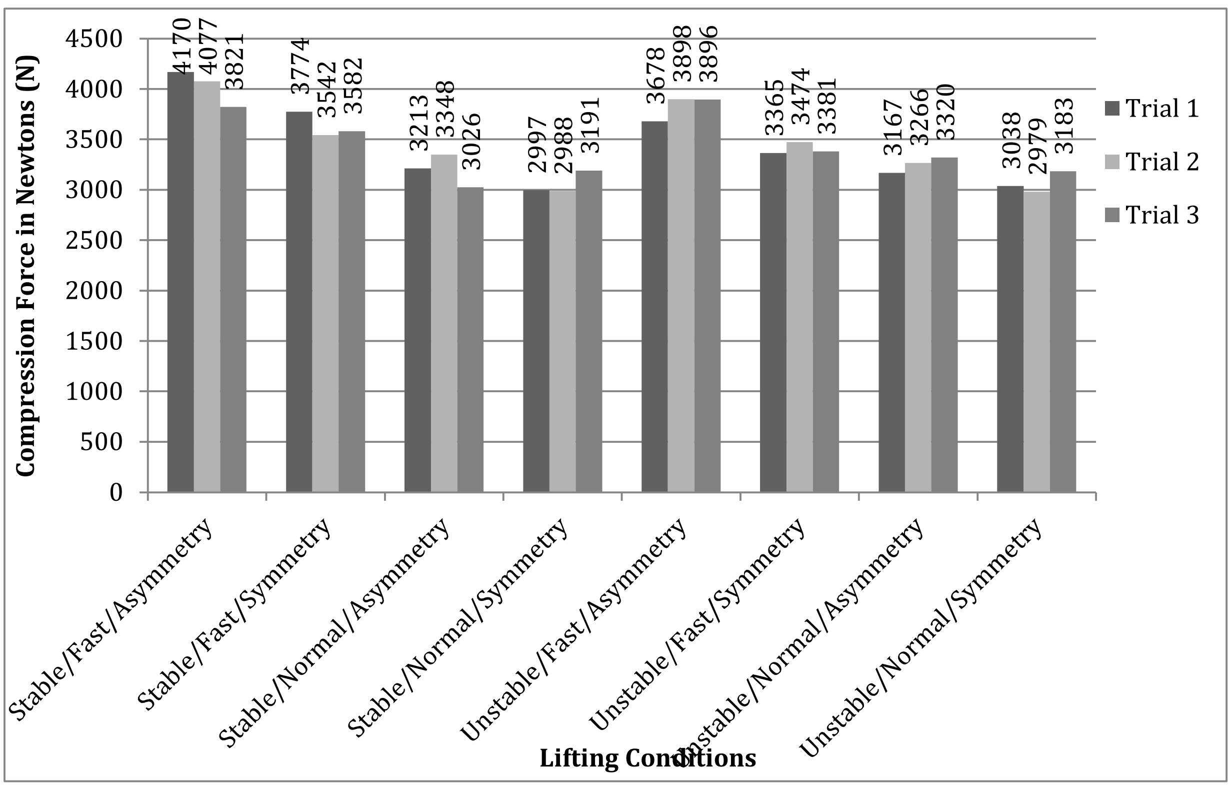

Figure 14: L5/S1 disc compressive forces (N) in eight lifting conditions and across three trials for a subject

When comparing peak compressive forces due to lifting styles (asymmetrical vs. symmetrical), asymmetrical lifting had a 10% greater force average compared to symmetrical lifting (Figure 15 and 16).

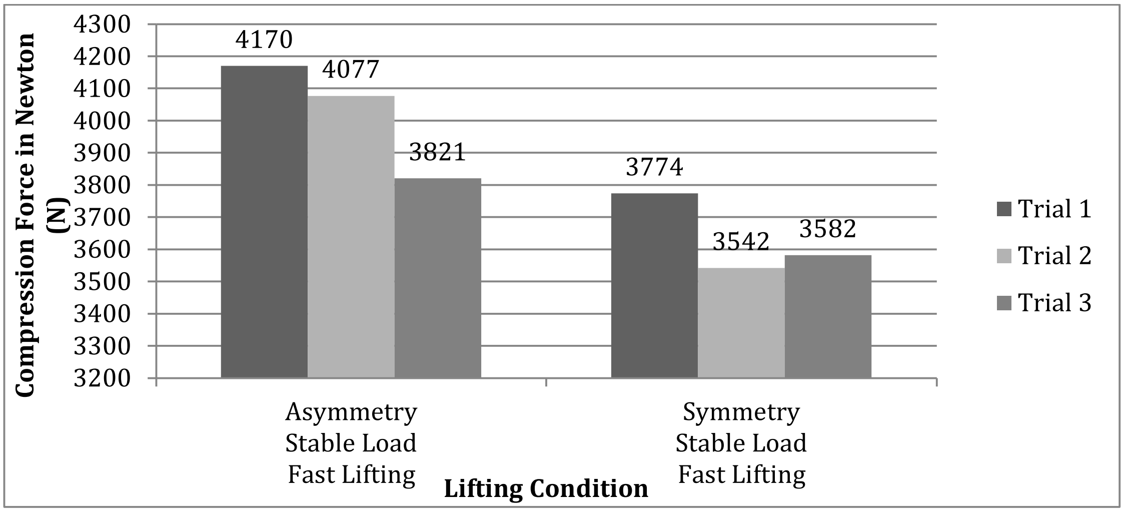

Figure 15: L5/S1 disc compressive forces (N) of a single subject lifting trials of a stable load at fast pace symmetrically vs. asymmetrically.

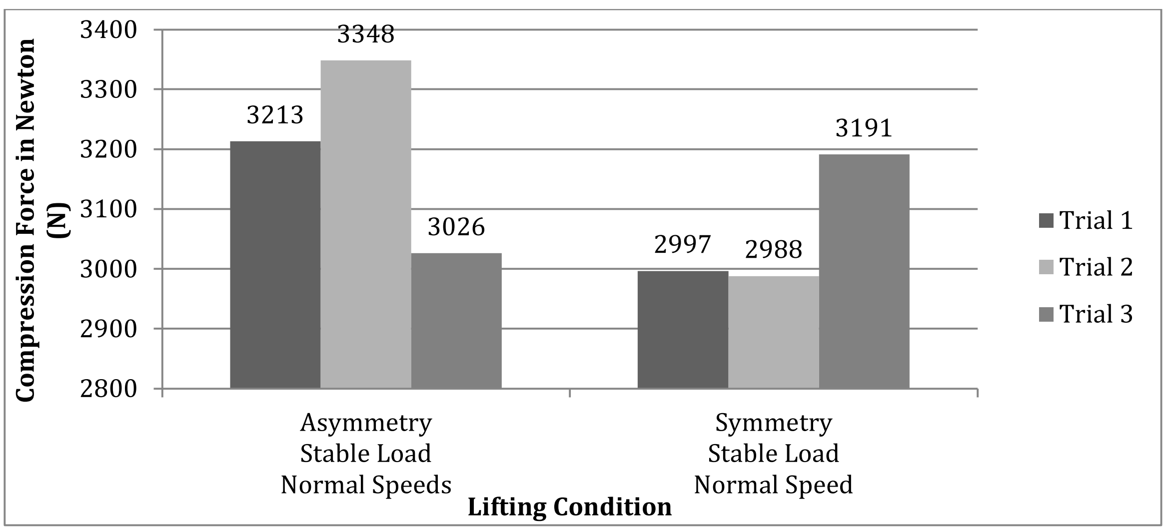

Figure 16: L5/S1 disc compressive forces (N) of a single subject lifting trials of a stable load at normal speeds symmetrically vs. asymmetrically.

Of the three lifting variables, the speed of lifting demonstrated the highest compression force. For asymmetric lifting, the faster lifting speed produced a compression force that was 25% greater (or 827N) than the compression force produced by normal speed lifting (Figure 16); and for symmetric lifting, the difference was 19% (Figure 17).

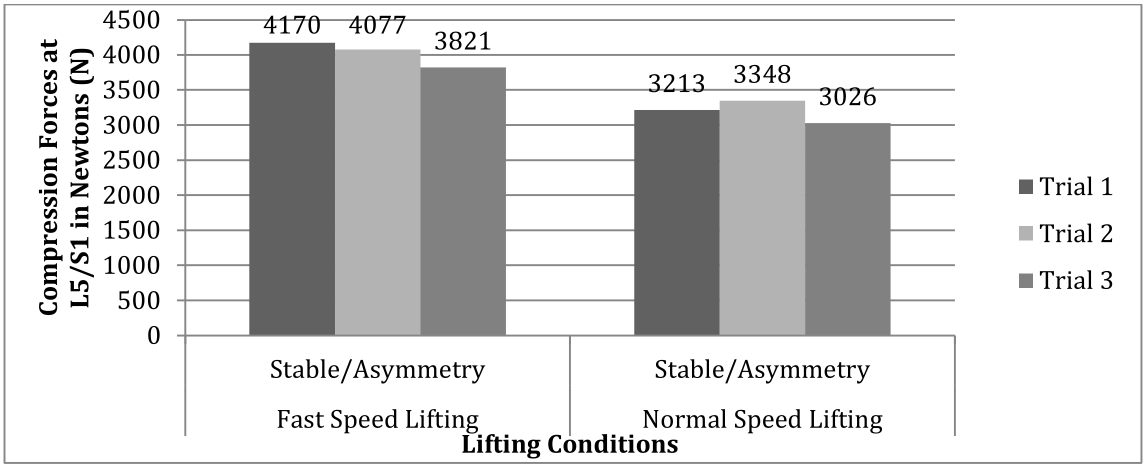

Figure 17: L5/S1 disc compressive forces (N) of a single subject with 3 lifting trials of a stable load asymmetrically at fast versus normal speeds

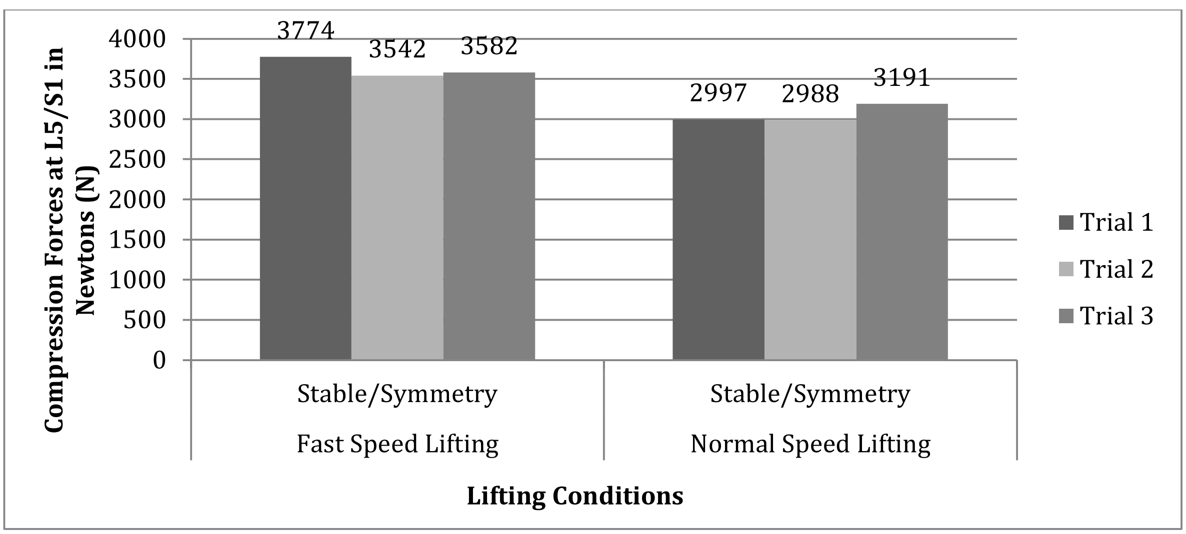

Figure 18: L5/S1 disc compressive forces (N) of a single subject with 3 lifting trials lifting a stable load symmetrically at fast versus normal speeds.

Of the three variables, load type did not show any noticeable compressive force difference between the two levels (stable vs unstable), with stable loads producing only a 5% difference (increase) compared to unstable load. This increase was contrary to expectation, but experimentation with a reasonable sample size is required to determine statistical significance.

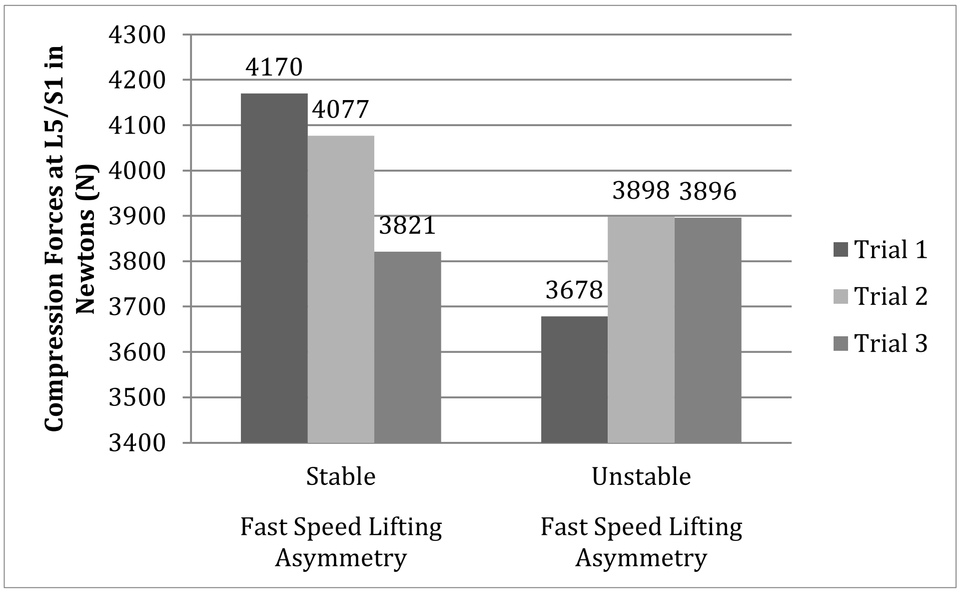

Figure 19: L5/S1 disc compressive forces (N) of a single subject with 3 lifting trials of a fast-asymmetrical lifting of an unstable and stable load

Asymmetrical lifting produced an 82 % higher torque moment compared to symmetric lifting. For asymmetrical lifting, the average torque moments at L5/S1 disc was greater by 34Nm (Figure 21) compared to symmetrical lifting. Asymmetrical lifting torsion moments at L5/S1 disc peaked at an average of 75Nm, while symmetrical lifting torsion moment average was only 41N (Figure 20 and 21).

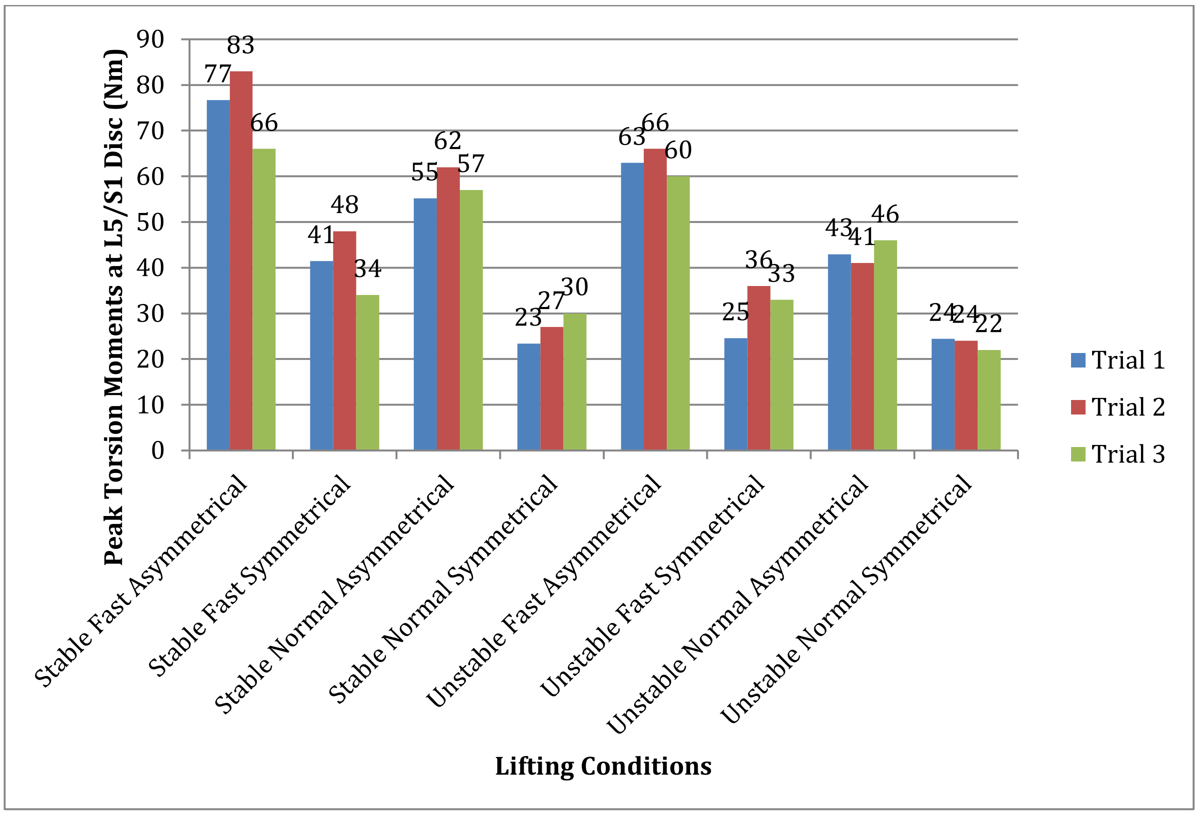

Figure 20: L5/S1 disc torsion moments (Nm) in eight lifting conditions and across three trials for a subject

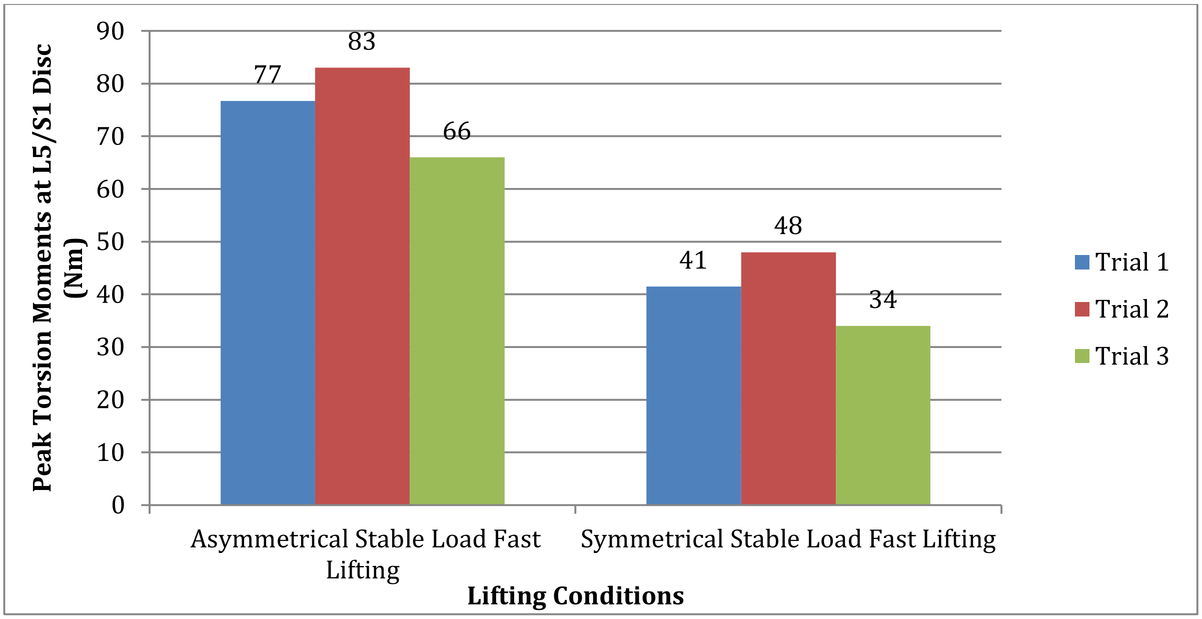

Figure 21: L5/S1 disc torsion moment (Nm) for fast lifting a stable load in an asymmetrical vs symmetrical style

Fast speed lifting also demonstrated higher peak torsion moments at L5/S1 disc, that was 30% (Figure 21) greater than normal speed lifting. For fast speed lifting, the average peak torsion moments at L5/S1 disc was greater by 17Nm compared to normal speed lifting. The torsion moments at L5/S1 disc for normal speed lifting averaged at, while it peaked at 75Nm speed lifting the same load (Figure 22).

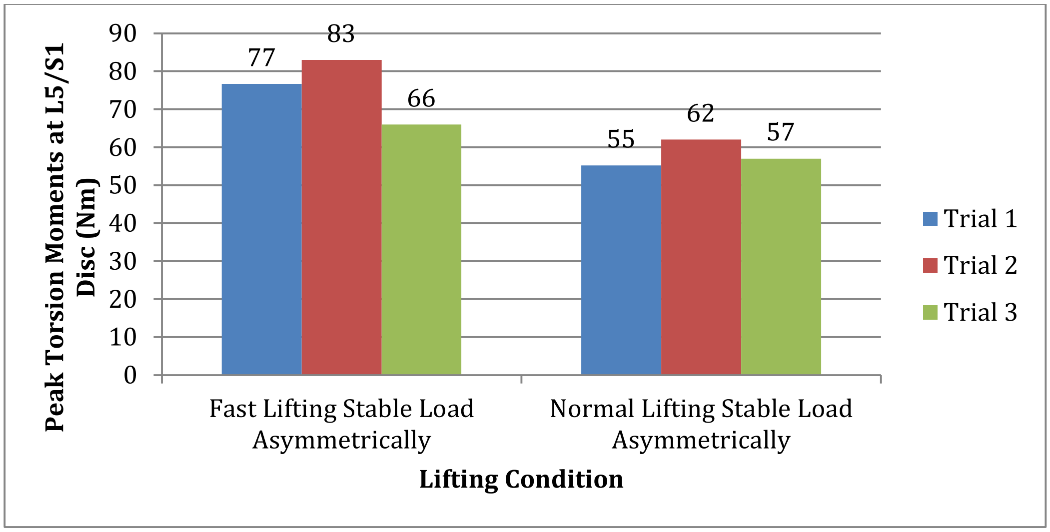

Figure 22: L5/S1 disc peak torsion moments (Nm) fast vs normal lifting a stable load asymmetrically

Lifting along the sagittal plane (symmetrical lifting) demonstrated high anterior-posterior shear forces (average=874N) at L5/S1 disc (Figures 23 and 24), demonstrating comparable forces for lifting asymmetrically (average=875N). Fast speed lifting demonstrated a greater anterior-posterior shear force at L5/S1 disc by an average of 30% compared to normal speed lifting (Figure 25). For fast speed lifting along the sagittal plane, the average anterior-posterior shear forces at L5/S1 disc was greater by 209N compared to normal speed lifting across the frontal plane. The anterior-posterior shear force at the L5/S1 disc lifting at normal speeds peaked at an average of 699N (Figure 25), while it peaked at 908N for fast speed lifting.

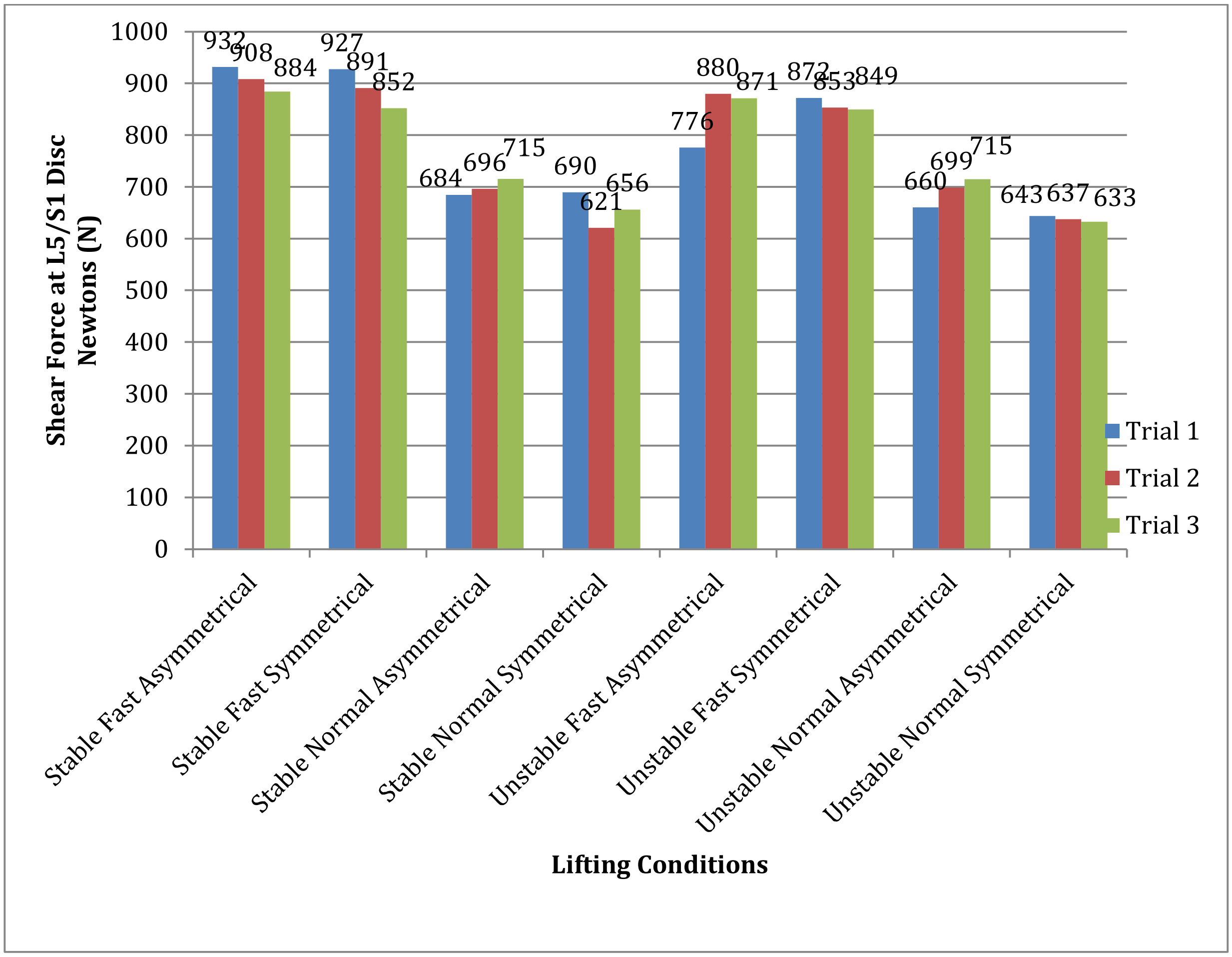

Figure 23: Anterior-Posterior L5/S1 disc shear Force (N) of eight lifting conditions across three trials for one subject

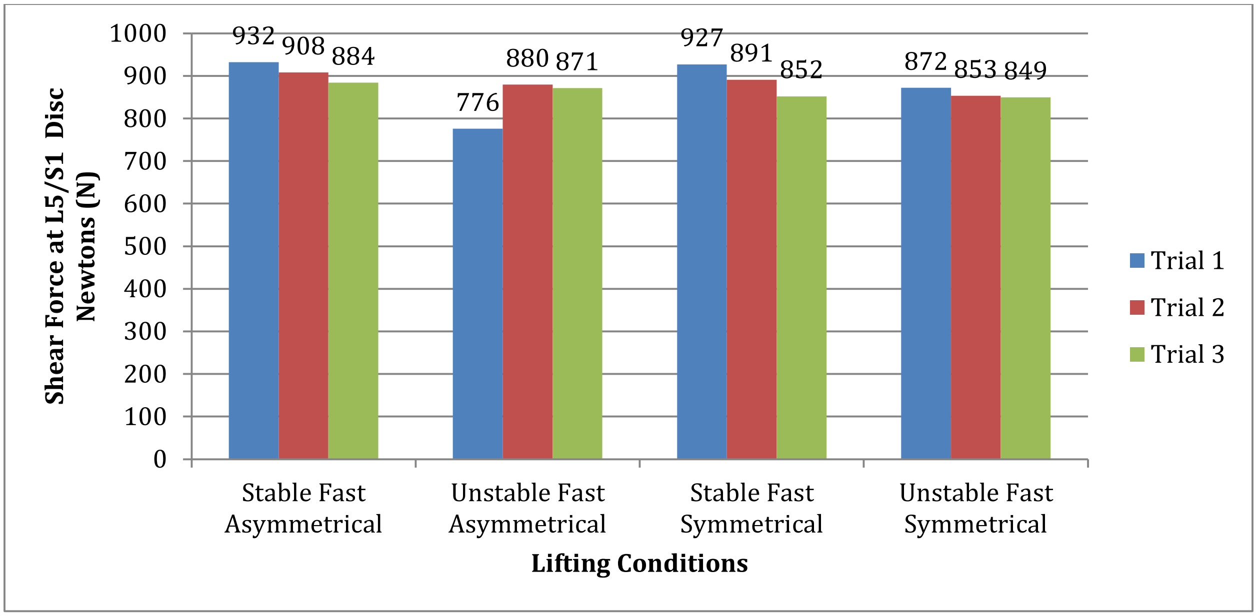

Figure 24: Anterior-Posterior L5/S1 disc Shear Force (N) of fast lifting a stable load symmetrical versus asymmetrical across three trials for one subject

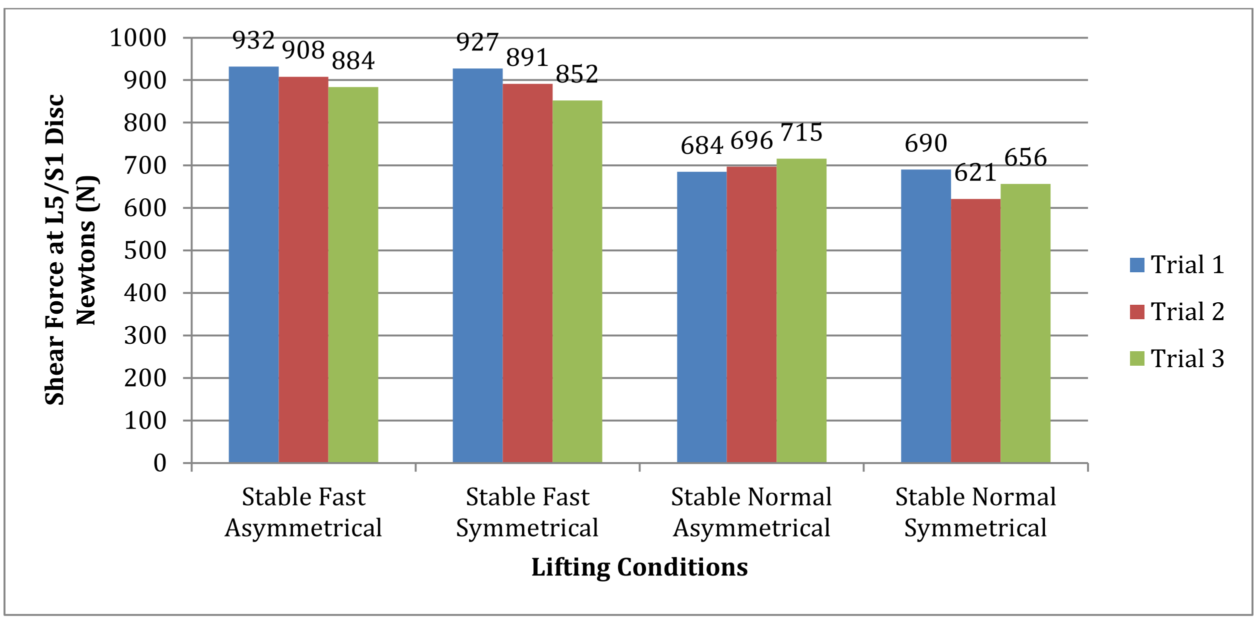

Figure 25: Anterior-Posterior L5/S1 disc shear force (N) of fast versus normal speed lifting a stable load symmetrical and asymmetrical across three trials for one subject

Lifting a load asymmetrically (90-degree angle of twist) increased the medial-lateral shear force at the L5/S1 disc at approximately 83% compared to lifting along the sagittal plane (symmetrical lifting) (Figures 26 and 27) – from 73N to 134N.

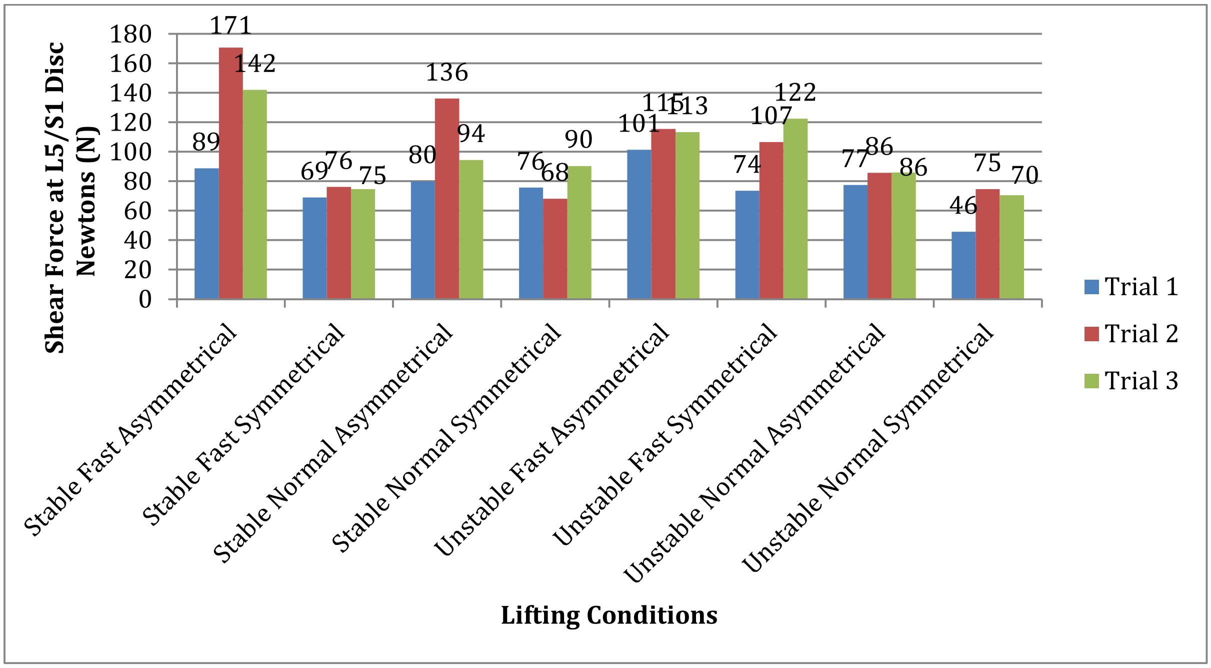

Figure 26: Medial-Lateral L5/S1 disc shear force (N) of eight lifting conditions across three trials for one subject

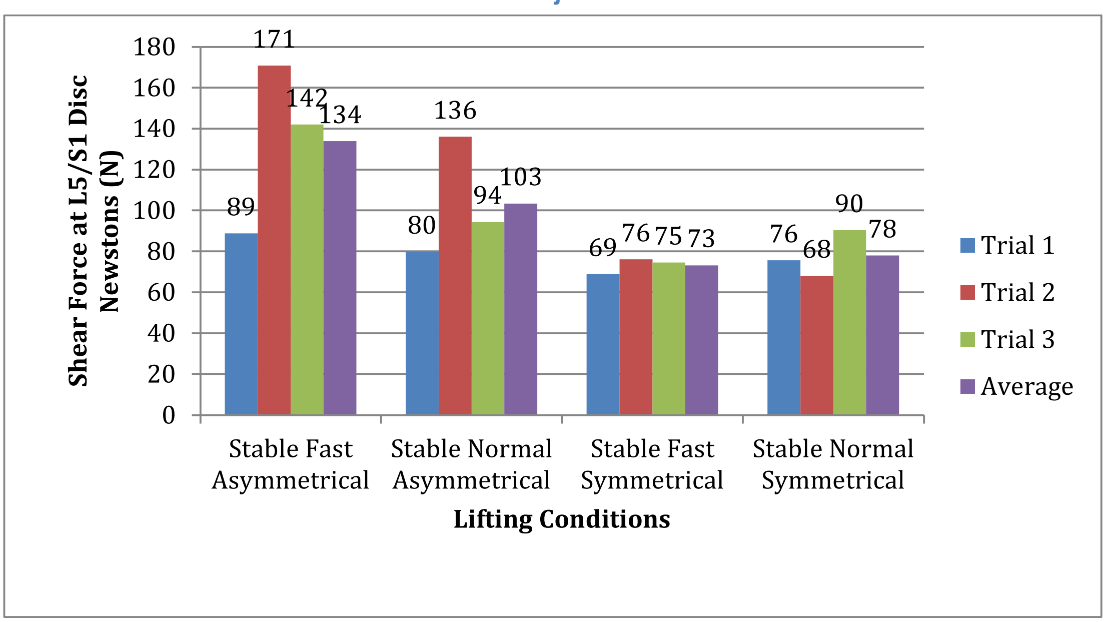

Figure 27: Medial-Lateral L5/S1 disc shear force (N) of fast lifting a stable load symmetrical versus asymmetrical across three trials for one subject

6. Discussion

In this study we developed a three-dimensional orthogonal coordinate system, with the origin located at the L5/S1 disc level (see Figure 28) and demonstrated the utility of this new technical coordinate system to track changes in forces acting on the lumbar vertebrate. The lumbar technical coordinate system was then used to compute compression and shear forces and torsional moments that can be used to investigate changes in lumbar forces and moments across a combination of lifting variables. We used a three-factor research design for two lifting types (Stable and Unstable), two speeds (Normal and Fast) and two different lifting styles (Symmetry and Asymmetry). To measure forces accurately and their effect on the trunk, lumbar and pelvis , we developed a customized lifting bin, partly filled with water, with two load cells (placed on the right- and left-hand side of the bin) to determine the effects of the shifting center of mass in X-Y-Z direction due to this liquid (Figure 30). Figure 29 shows angles for the pelvic, lumbar and trunk segments relative to the global coordinate system in a sample trial of fast asymmetrical lifting of an unstable load. As shown in Figure 29 we were able to derive the lumbar motion separately using the trunk and pelvic coordinate systems.

Figure 28 illustrates changes in the angles from a typical trial of fast asymmetric lifting with an unstable load. Anterior and medial/lateral shear forces are shown in the top graph, compression forces in the middle graph, and torsion moment about the L5/S1 disc vertical axis is shown in the bottom graph. Dashed vertical lines relate the body position of the lifter at the peak forces and moments. Peak anterior shear of 930.67 N occurs at time = 0.28 s after the start of the lift, and peak compression force of

4,169.81N occurs at time = 0.29 s (middle graph). Peak torsion moment about the vertical axis at the L5/S1 disc, of 76.69 Nm, occurs at tme = 1.42 s

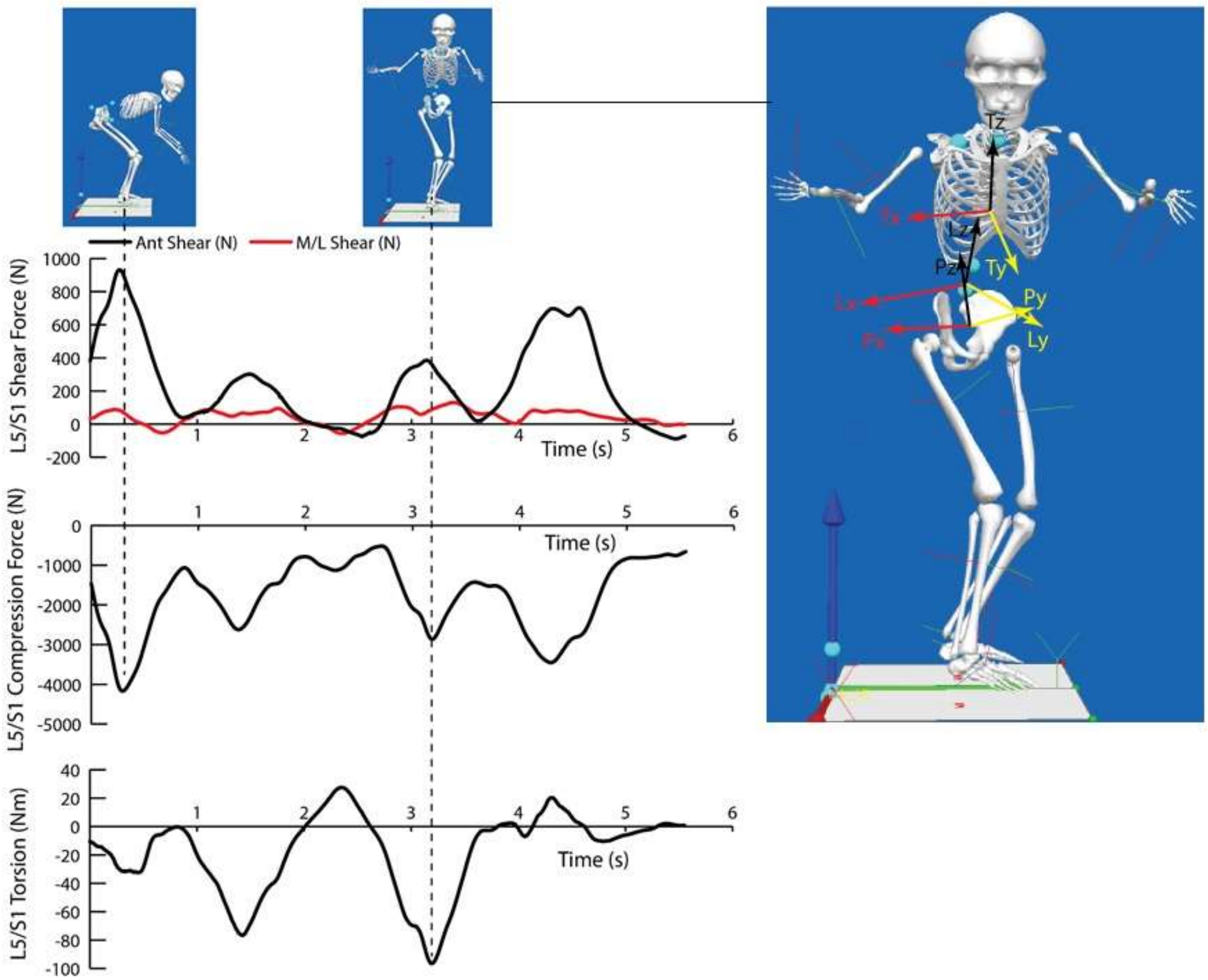

Figure 28: Asymmetrical trial of an unstable load with anterior-posterior, medial-lateral shear forces (N) and torsion moments (Nm) at L5/S1 disc from start of lifting

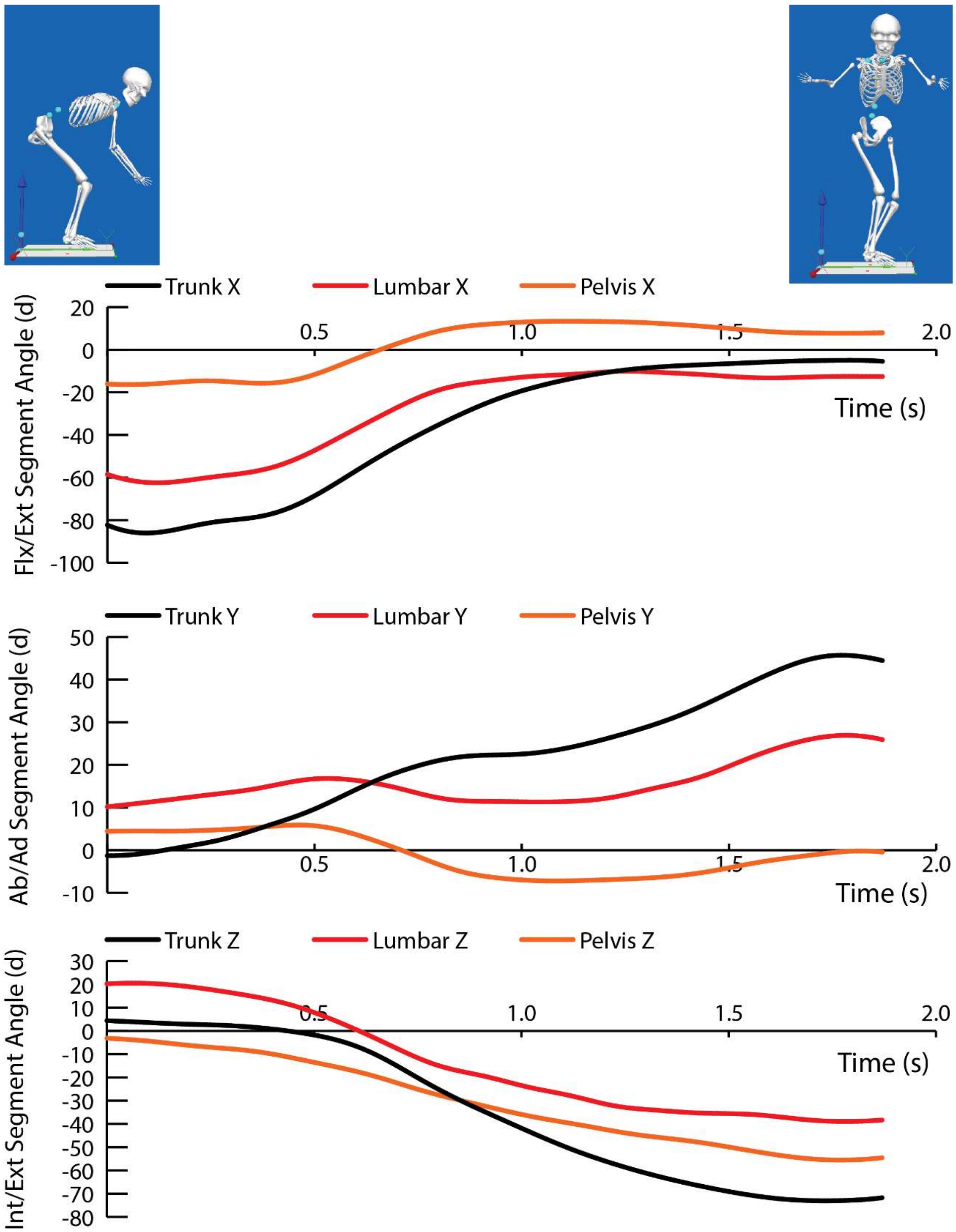

Figure 29: Captured snapshot of subject's pelvic , thoracic, and lumbar degrees motion of angles (d) during lifting start from lifting time (S)

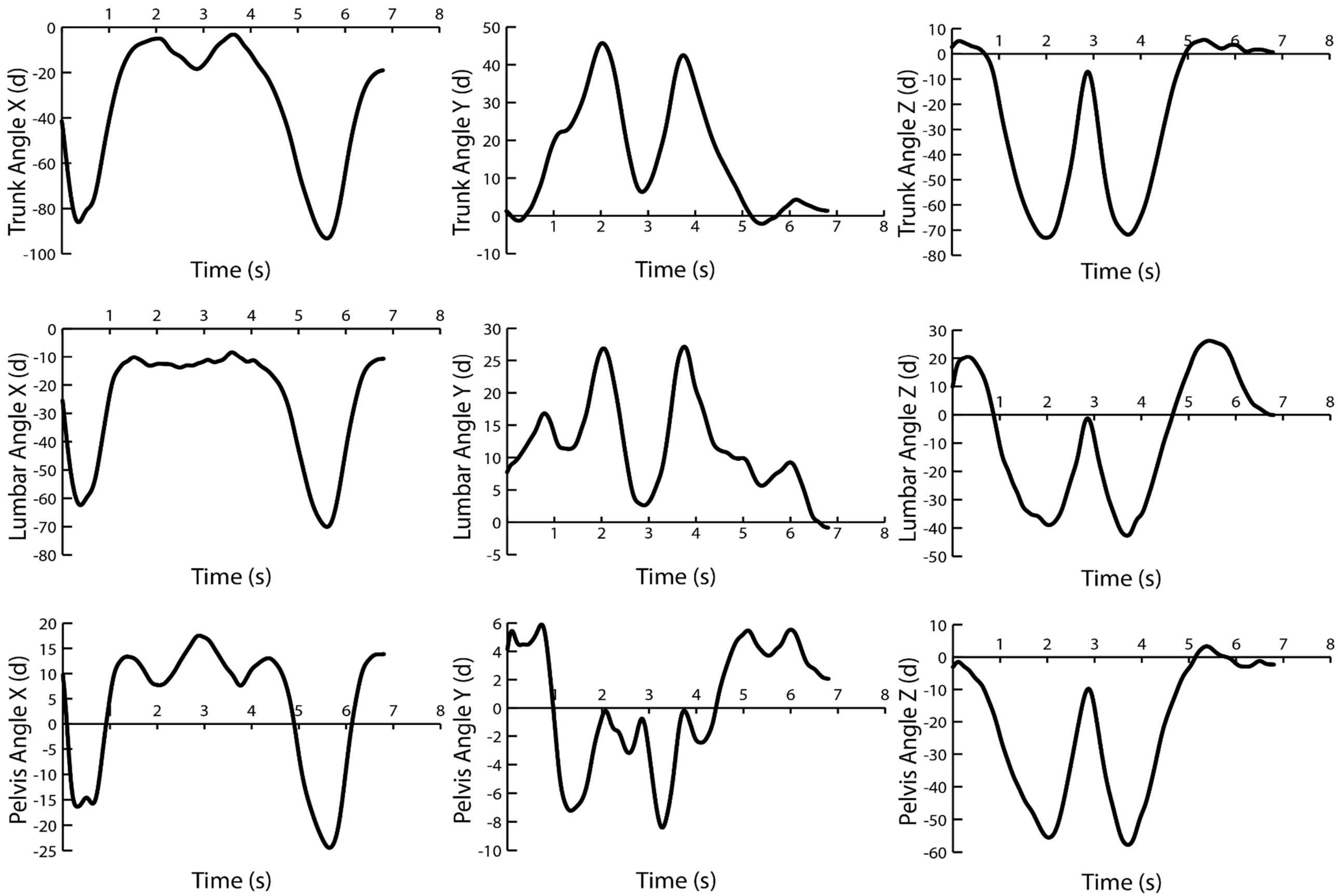

Figure 30: Comparison of Thoracic, Pelvic and Lumbar Motion Segment Angles (d) in three dimensions X-Y-Z, starting from time of lifting (S).

Table 3 presents a comparison between the results of this study (compression and shear forces, and torque moments) at the L5/S1 disc to other related published papers. Van Dieen et al. (2001;2003) in two studies tested lifting an unstable load (liquid water) versus a solid stable load. Unstable load lifting presented a 9% compression force increase in their first study (Van Dieen et al., 2001) compared to stable loads, and a 15% increase on their second study (Van Dieen et al., 2001). Their study used EMG data to estimate compression forces. However, our method analyzed body joint angles, velocities, accelerations (using a visual 3D Vicon camera system), and forces from force plates and load cells, at the L5/S1 disc technical coordinate system. The new method demonstrated a comparable 10% compressive force increase when lifting fast at speeds.

In this study, the average compression force values were 3430N, slightly higher than the NIOSH lower limit of 3,400N at the L5/S1 disc for lifting safe loads, but less than the (dangerous) maximum permissible limit of 6,400N (NIOSH, 1981).

Like previous studies (Table 3), this study has also demonstrated that fast speed lifting and asymmetrically lifting had produced higher compression (4023N), anterior-posterior shear (852N) and torsion moments (74.5N) at L5/S1 disc.

The anterior-posterior shear forces at the lumbar spine are not as well documented as the compression forces. McGill et al. (1988) recommended the maximum permissible limit anterior-posterior shear force to be 1,000N. Other researchers (Gallagher and Marras, 2012) suggested a 1,000N anterior-posterior shear for infrequent loading, with less than 100 loads per day. They suggested less than a

700N of anterior-posterior shear force limit for frequent loading up to 1,000N limit for infrequent loading. Tests on working age cadaver specimens found the anterior lumbar spine to fail at approximately 1,200N (Begeman et al., 1994). The anterior-posterior shear forces for this study ranged between 633N to 852N, well below the upper limits stated in the literature.

In a study by McGill, 1991, using EMG activity of the trunk muscles, the torsion moments at the L5/S1 disc ranged between 25N and 102N. Marras et al. (1995) found between 52N and 90N. In this study, the torsional moments were comparable: 23-75N.

Previous studies have identified asymmetrical lifting, fast speed lifting and load stability to be significant factors in lumbar spine loading (Van Dieen et al., 2001; Davis K et al., 1998; Marras et al., 2000; McGill 1991; Marras et al., 1995; Marras et al., 2012). The analysis in this study focused primarily upon factors that resulted in significant differences in spinal loading at L5/S1 disc. Using state-of-art technology, this study provides a new methodology to estimate the effects of those significant MMH lifting factors. None of the previous research attempted to estimate torsion moments, compression, and shear forces at L5/S1 disc during dynamic asymmetrical, fast lifting of an unstable load. In general, this summary indicated that our new method for measuring compression and shear forces along with torsion moments at L5/S1 disc using the three-factor research design (load type, lifting speed and lifting style) have resulted in forces and moments that fall within range of previous studies

This study utilized EMG electrodes to provide a comprehensive approach in estimating compression and shear forces along with torque moments at L5/S1 disc.

However, significant difficulties were detected. Electromyographic electrodes would fall off during testing because subjects had numerous reflective markers (97) taped on them. While this study failed to collect EMG data, future studies should incorporate EMG analysis for a more accurate and reliable data analysis.

Table 3: A comparison between the data results of this study (compression and shear forces (N), torque moments(Nm)) at L5/S1 disc to other related published papers

	Force Estimation Methodology	Lifting Conditions	Compression Force (N)	Anterior-Posterior Shear Force (N)	Moment Torsion (Nm)
This Study	Three Dimensional Biomechanical using Technical Coordinate System of L5/S1 disc	Symmetrical and. Asymmetrical Fast and Normal Speed Lifting Stable and Unstable Loads	3059-4023	633-852	23-74.5
Van Dieen et al., (2001)	EMG Three-dimensional linked segment biomechanical model using	Symmetrical As fast as possible Stable and unstable loads	2000-8000	N/a	N/a
Davis K et al., (1998)	Electromyographic Activity of Muscles and 3D electro-goniometer Lumbar Motion Monitor (LMM)	Asymmetrical Stable Load	2620-3269	680-815	N/A
Marras et al., (2000)	Three-dimensional EMG-assisted biomechanical Model and LMM	Symmetrical and Asymmetrical Stable Load	3148-4257	598-816	44-73
McGill (1991)	Electromyographic Activity of Muscles	Symmetrical and Asymmetrical	N/A	N/A	25-102
Marras et al. (1995)	Electromyographic Activity of Muscles	Asymmetrical	N/A	N/A	52-90
Marras et al. (2012)	Literature review paper for maximum permissible limit of shear forces	Papers reviews mainly used EMG	N/A	760N Frequent Loading 1140N Infrequent Loading	N/A

N/A: No values were reported

7. Conclusions

In this study we presented a new methodology to estimate forces at a more precise location of L5/S1 disc. We created a technical lumbar coordinate system to define the precise location of L5/S1 disc. The lifting experiment process involved two different load types (Stable and Unstable), two speeds (Normal and Fast) and two different lifting styles (Symmetry and Asymmetry). We customized a lifting bin with two load cells to determine the effects of the unstable load (liquid) shifting center of mass. We were able to determine the actual peak compression, shear, and torque moment at the estimated location of L5/S1 disc.

The results from a single subject are not conclusive, however, this methodology provided a new technique to collect a comprehensive force analysis (compression, shear, and torsion moments) acting on a lifter at L5/S1 disc. None of the previous studies presented in the literature above attempted to test unstable load lifting asymmetrically along with speed lifting. Using the customized bin with load cells provides us a more acting on a lifter method to detect the shifting center of mass of the unstable load (liquid water).

Finally, potential study limitations should be acknowledged. Only one of the subject's data was sufficient to extract forces and moments at L5/S1 disc, out of the three subjects we tested. Furthermore, we were not able to process the EMG signals collected from that subject. Thus, the estimated torsion moments at L5/S1 disc are slightly underestimated due to absence of actual muscle contractions of the latissimus dorsi and external/internal oblique muscles.

This study has opened the door to many future studies relevant to determining the effects of lifting load types (stable and unstable) lifting style (symmetrical and asymmetrical), and lifting speed (fast, slow, and normal). For instance, further research using this methodology can help determine if unstable loads have a greater negative impact on the lumbar spine compared to stable loads.

Appendix A:

Tyler’s Full Body Marker Abbreviations

Table 3: Tyler's Full Body Marker Set

Upper Body	
Head	
THead	Top of Head
PHead	Posterior Head
C7	Cervical Vertebrae 7
Ahead	Anterior Head
RHead	Right Head
LHead	Left Head
RNeck	Right Neck
LNeck	Left Neck
Trunk	
RAC	Right Acromial Joint
LAC	Left Acromial Joint
CLAV	Clavicle
STERN	Sternum
T2	Thoracic Vertebrae 2
T8	Thoracic Vertebrae 8
RLT	Right Lower Trunk (lowest floating rib on right side)
LLT	Left Lower Trunk (lowest floating rib on left side)
RUT	Right Upper Trunk (in line with the base of the sternum)
LUT	Left Upper Trunk (in line with the base of the sternum)
Lumbar	
L1	L1 Vertebra
L3	L3 Vertebra
L5	L5 Vertebra

Table 3: Tyler's Full Body Marker Set

Right Upper Arm	
RADL	Right Anterior Deltoid
RPDL	Right Posterior Deltoid
RUA1	Right Upper Arm 1 (cluster)
RUA2	Right Upper Arm 2 (cluster)
RUA3	Right Upper Arm 3 (cluster)
RUA4	Right Upper Arm 4 (cluster)
RLEL	Right Lateral Elbow
Right Forearm	
RMEL	Right Medial Elbow
RFA1	Right Forearm 1 (cluster)
RFA2	Right Forearm 2 (cluster)
RFA3	Right Forearm 3 (cluster)
RFA4	Right Forearm 4 (cluster)
RWRR	Right Wrist Radial
RWRU	Right Wrist Ulnar
Right Hand	
RHR	Right Hand Radial
RHM	Right Hand Middle
RHU	Right Hand Ulnar
Left Upper Arm	
LADL	Left Anterior Deltoid
LPDL	Left Posterior Deltoid
LUA1	Left Upper Arm 1 (cluster)
LUA2	Left Upper Arm 2 (cluster)
LUA3	Left Upper Arm 3 (cluster)
LUA4	Left Upper Arm 4 (cluster)
LLEL	Left Lateral Elbow

Table 3: Tyler's Full Body Marker Set

Left Forearm	
LMEL	Left Medial Elbow
LFA1	Left Forearm 1 (cluster)
LFA2	Left Forearm 2 (cluster)
LFA3	Left Forearm 3 (cluster)
LFA4	Left Forearm 4 (cluster)
LWRR	Left Wrist Radial
LWRU	Left Wrist Ulnar
Left Hand	
LHR	Left Hand Radial
LHM	Left Hand Middle
LHU	Left Hand Ulnar
Lower Body	
Pelvis	
RPP	Right Pelvis Peak
LPP	Left Pelvis Peak
RPS	Right PSIS (Posterior Superior Iliac Spine)
LPS	Left PSIS (Posterior Superior Iliac Spine)
Right Thigh	
RHP	Right Hip (Greater Trochanter)
RTH1	Right Thigh 1 (cluster)
RTH2	Right Thigh 2 (cluster)
RTH3	Right Thigh 3 (cluster)
RTH4	Right Thigh 4 (cluster)
RLK	Right Lateral Knee
RMK	Right Medial Knee
Right Shank (lower leg)	
RTT	Right Tibial Tuberosity

Table 3: Tyler's Full Body Marker Set

RSK1	Right Shank 1 (cluster)
RSK2	Right Shank 2 (cluster)
RSK3	Right Shank 3 (cluster)
RSK4	Right Shank 4 (cluster)
RLA	Right Lateral Ankle
RMA	Right Medial Ankle
Right Foot	
R1MH	Right 1st Metatarsal Head
RToe	Right Toe (in between 2nd and 3rd metatarsal head)
R5MH	Right 5th Metatarsal Head
R5MB	Right 5th Metatarsal Base
RHL	Right Heel
Left Thigh	
LHP	Left Hip (Greater Trochanter)
LTH1	Left Thigh 1 (cluster)
LTH2	Left Thigh 2 (cluster)
LTH3	Left Thigh 3 (cluster)
LTH4	Left Thigh 4 (cluster)
LLK	Left Lateral Knee
LMK	Left Medial Knee
Left Shank (lower leg)	
LTT	Left Tibial Tuberosity
LSK1	Left Shank 1 (cluster)
LSK2	Left Shank 2 (cluster)
LSK3	Left Shank 3 (cluster)
LSK4	Left Shank 4 (cluster)
LLA	Left Lateral Ankle

Table 3: Tyler's Full Body Marker Set

LMA	Left Medial Ankle
Left Foot	
L1MH	Left 1st Metatarsal Head
LToe	Left Toe (in between 2nd and 3rd metatarsal head)
L5MH	Left 5th Metatarsal Head
L5MB	Left 5th Metatarsal Base
LHL	Left Heel

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