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Studies of compressive forces on L5/S1 during dynamic manual lifting

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Studies of compressive forces on L5/S1 during dynamic manual lifting

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For the Major Program

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

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ABSTRACT

The lifting task is the major activity contributing to the risk of lower back injury. The purpose of this study was to examine the mechanical stresses on the lower back during manual lifting and to analyze how people adapt to the stresses. This purpose was accomplished by approaching the mechanical stresses on the lower back from three different perspectives. The effects of manual lifting tasks were measured as: the fatigued condition, the knowledge of load and the previous lifts, and effort in lifting over a constraining barrier. A biomechanical model (linked segment and rigid body) was applied to estimate the peak compressive forces at the L5/S1 (lower back) joint during the manual lifting tasks. Three-dimensional locations of joint markers were digitized using a Peak Video and Analog Motion Measurement System. A strain-gage force platform was used to determine the location, direction and magnitude of external ground reaction forces acting on the feet. The peak compressive forces at the lower back were insignificantly different when individuals were fatigued. Some individuals increased and others decreased when they were fatigued. Individuals tended to lift faster and bring the load closer to the body when fatigued. Individuals applied different techniques of lifting when they were fatigued. Individuals lifted and adapted differently during the subsequent lifts when lifting known or unknown loads of different masses. Lifting under an unknown condition generated greater stresses on the lower back, especially when the load was light. When lifting a light load, individuals adapted to the lifts by bringing the load closer to the body, which reduced the stresses on the lower back. However, when the load was heavy, individuals adapted the lifts by changing the lifting technique, which did not reduce the stresses on the lower back. Lifting over a constraint barrier developed greater forces on the lower back. Individuals required more flexing of the lower back and less bending of the knees when lifting over the barrier. Individuals adapt to their internal or external lifting environment. These adaptations can increase or decrease the peak compressive forces.

CHAPTER 1: DEVELOPMENT OF THE PROBLEM

Introduction

Manual material handling (MMH) tasks are required in many jobs and activities, such as loading and unloading of boxes/cartons, removing materials from a conveyer belt, or stacking items in a warehouse. Consequently many workers may be at risk of suffering work-related injuries performing those activities. The National Safety Council reported that more workers injured their backs than any other body part from 1987 to 1994 (Mital et al., 1997). In addition, the National Institute of Occupational Safety and Health (NIOSH) estimated that the total cost of back injuries in the United States in 1991 was between 50 and 100 billion dollars (Mital et al., 1997). According to these facts, back injuries associated with manual material handling tasks have been a serious problem for both the individual and the national economy.

There are many different types of the MMH tasks, such as lifting, holding, carrying, pushing, and pulling. Nearly 50% of such injuries occur while lifting objects, while only 9% occur while holding, wielding, throwing, or carrying objects (Klein, Roger, Jensen, and Sanderson, 1984). Therefore, lifting is considered the most common MMH task associated with the occurrence of low back injuries. This is why many research studies have concentrated on the manual lifting tasks to develop knowledge of injury prevention in the low back.

One common aspect in many studies of lifting tasks is the comparison of lifting techniques. There are two different techniques that are often analyzed: squat (straight-back, bent-knee) and stoop (bent-back, straight-knee). The squat technique produces less stress in the lower back than the stoop (Andersson and Chaffin, 1986). However, the squat technique was the most demanding technique, requiring the highest oxygen consumption and inspiratory ventilation volume, of the two techniques (Kumar, 1984). So, in reality, people do not use these two extreme techniques, they instead use a combination of both techniques, the free-style technique.

Many studies examined the lifting task by using a biomechanical approach, in which stresses incurred in the body while lifting are estimated. Focus is primarily on the lower back, especially the L5/S1 disc (Buseck et al., 1988, Busch-Joseph et al., 1988). These studies used a linked-segment and rigid body model to estimate the flexion-extension forces and moments that occurred in the lower back during lifting trials. The biomechanical models that have been used in the study of manual lifting tasks require kinematic and kinetic data collected during the movement. To collect the kinematic data, some researchers placed a variety of markers on the body, i.e., joint centers, and detected the positions of each marker by using video-camera systems (Patterson et al., 1987, Gagnon et al., 1996). For the collecting of kinetic data, some researchers used the force platform system to measure the ground reaction forces of the body that developed between the subject's feet while standing on the force platform. This ground reaction force provided data to calculate the forces and moments of each joint in the body (Schipplein et al., 1990). To increase the accuracy of biomechanical models, subject anthropometry data are required. This data provides the information needed on the human body, such as the position of joint centers, mass, center of mass, radii of gyration, and segment length for each body segment. Some anthropometry data have been obtained by using different measurement techniques, such as gamma ray (Zatsiorsky and Seluyanow, 1983), and computed tomography (Pearsal et al., 1996).

Lifting objects is certainly a physically exerting task, especially in lifting tasks that must be done repetitively. Repetitive lifting has been shown to be a risk factor for the development of low back injury (Frymoyer et al., 1983) and caused a rapid development of back extensor muscle fatigue (Petrofsky and Lind, 1978, Potvin and Norman, 1993). The fatigued condition as related to the particular risks associated with the injury of lower back during lifting has been investigated. Chen (2000) found that the peak compressive forces on the L5/S1 disc were increased when localized arm fatigue was generated. However, Dolan and Adams (1998) discovered that peak spinal compression at lower back was decreased after the repetitive lifting task was performed. These previous studies have shown different results of the compressive forces at lower back when fatigue condition was reached. The first study applied a fatigue protocol of particular parts of the body that emphasized control. The second study used multiple lifts to fatigue the entire body, however, the fatigue condition of the subjects was not until volitional failure. The subjects were asked to lift 100 times, so the magnitude of fatigue was probably different between subjects. Thus, a further study of the mechanical stresses on the lower back as the effect of fatigue, which was generated from the repetitive lifts, was justified.

The daily situation of workers like refuse collectors, luggage dispatchers and movers is fundamentally different from the situation in which an unsuitable preparation is achieved by giving false expectations. Epidemiological studies have shown that workers exerting sudden unexpected maximal efforts are particularly vulnerable to low back disorders (Magora, 1973). Some experimental lifting studies have investigated the effect of load knowledge on the lower back loading. Most of the studies found that lifting an object with versus without load knowledge resulted in an increased lumbo-sacral loading in the latter condition (Patterson et al., 1987, Butler et al., 1993, Commissaris and Toussaint, 1997, de Looze et al., 2000). However, one study showed that the absence of load knowledge, particularly the underestimation of the object mass did not lead to an increase in low back loading (der Burg and van Dieen, 2001). These previous studies have shown different results of the stresses occurred at lower back as the effect of knowledge of the load. None of these studies investigated the effect of previous lifts to the subsequent lifts either under the known or unknown condition. The investigation of how people adapted the previous lift during the subsequent lifts could be a good prevention of the injury that possible happened during lifting tasks. Thus, a further study of the mechanical stresses on the lower back reflecting the effect of load knowledge and the effect of the previous lifts was performed.

Many manual material handling (MMH) tasks require individuals to lift from, and lower into, industrial bins. These bins or barriers can serve to constrain the flexion of the knee and this can have implications for spine flexion during load handling. However, only one study (McKean and Potvin,

2001) showed that the constraint barrier appeared to increase the risk of injury during lifting and lowering by increasing horizontal reach, trunk flexion and magnitude of muscle activity (EMG). This previous study did not calculate the stresses on the lower back (i.e., moments, compressive forces) during lifting under the constraint barrier condition. The previous study also only compared lifting tasks over a single constraining barrier (120% of knee height) to the non-constrained condition (freestyle). Thus, a further study, which estimated the mechanical stresses on the lower back as the response of different heights of constraining barrier, was performed.

Statement of the problem

Excessive amounts of stress on the lower back disc during manual work may cause serious problems in the lower back. Decreasing the magnitude of these stresses or preventing the effects of the mechanical stresses may reduce the injury potential. The mechanisms used to reduce these stresses are largely unknown. By examining the effects of mechanical stresses and the mechanisms involved in producing the amount of stresses, it may be possible to more accurately predict mechanisms that will lead to a reduced injury potential.

The purpose of this research was to examine the mechanical stresses at the lower back and to analyze how people adapted the stresses during manual lifting. Considering the mechanical stresses of the lower back enables preventing lower back injuries while studying different aspects (conditions) of the manual lifting tasks. This purpose was accomplished by approaching the mechanical stresses on the lower back from three perspectives. The first was to examine the stresses at the lower back and to analyze how individuals adapted the fatigue condition during continuously manual lifting. The second was to determine the stresses at the lower back as the effects of previous lifts and the knowledge of the load during the lifts. It was investigated how individuals adapted during the subsequent lifts either when they had knowledge of the load or when they did not. The lifts were performed under two load magnitudes (light and heavy) and two levels of load knowledge (known and unknown). The purpose of the third portion of this study was to determine stresses at the lower back and to analyze how individuals adapted during lifting a load over the constraining barrier with varying level of heights. A bar as the constraining barrier was placed between the load and subject during the lifting task.

Limitations

The following limitations applied to this study:

The subjects were selected from a somewhat heterogeneous population of Iowa State University students. In order to establish some amount of control over the environment and to allow instrumentation of the subjects, all lifting trials took place in the laboratory environment. It is acknowledged that this is not the actual manual material-handling environment and the results could have been affected.

Assumptions

The following assumptions were made in this study:

1.The biological signals were sampled at an adequate frequency (kinematics: 120 Hz, ground reaction forces: 120 Hz, electromyography: 1000 Hz).

2.The sample of subjects adequately represents a typical working population.

3. Marker placement and marker movement effects relative to the joint centers are minimal and will not affect the results and conclusions.

4. The assumptions of a segmented rigid body hold. In particular that the body can be described by segment point masses that have fixed, hinged joints, constant segment moments of inertia and constant segment lengths. The model also assumes minimal co-contraction of antagonistic muscles that cross L5/S1 and minimal contribution of intrerabdominal pressure of the extensor moment.

CHAPTER 2: REVIEW OF LITERATURE

Introduction

A wide variety of jobs require manual material handling (MMH), and consequently many workers may be at risk of suffering job-related injuries performing those activities. The National Safety Council published data that indicated that from 1987 to 1994 more workers had injuries to the back than any other body parts (Mital et al., 1997). Back injuries are not only prevalent, but they can also be debilitating. In 1985, The National Center for Health Statistics reported that back impairment was the most common cause of chronic activity limitation in people under age 45 and ranked third in those aged 45-64 (Khalil et al., 1993). Compounding the problems of frequency and severity is the great economic burden to the individual and the economy associated with back injuries. The National Institute of Occupational Safety and Health (NIOSH) estimated that the total cost of low back injuries for the United States in 1991 was between 50 and 100 billion dollars (Mital et al., 1997). Furthermore, the low back pain is considered to be the most expensive health care problem in the 30-50 age group (Khalil et al., 1993).

Low back injuries

The most common cause of low back injury is muscle or ligament strain, which is often caused by excessive lifting, off-balance motion, or fatigue. The injuries may result in a reduction in range of motion due to stiffness, weakness, and painful muscular spasms. Another common source of low back injury is the degeneration of the intervertebral discs, which is most likely caused by a combination of repeated overexertion and physiological changes due to aging. As a result, tears develop in the annulus fibrosus, the nucleus pulposus in a disc segment propagates into the annulus, and the intervertebral disc begins to bulge. A herniated disc that presses on a nerve root will cause back pain and may result in sciatica, which is pain in the back of the leg and foot associated with the compressed nerve (Breakstone, 1992). The nucleus pulposus may also propagate toward the end plates, causing weakening of the junction between intervertebral disc and the vertebra (Andersson, 1993). Injuries to the fourth lumbar/fifth lumbar (L4/L5) and the fifth lumbar first sacral (L5/S1) intervertebral discs account for 95% of all disk injuries in the lumbar vertebrae (Gracovetsky, 1990). Intervertebral disc degeneration can lead to more serious injury in the lumbar vertebrae, such as damage to the articular bony processes. Degeneration at these apophyseal joints may cause a condition called spondylolisthesis, in which vertebrae shift forward and out of alignment with the rest of the lumbar vertebrae.

After the onset of low back injury, 70% of all people can expect to get better within three weeks without specific treatment, and 90% will be free of discomfort in eight weeks (Breaksone, 1992). However, low back injuries have a high rate of recurrence and approximately 60% of those who have recovered will suffer again from pain within the first year after injury. Treatment of low back injuries ranges from restricting activities, taking medication, and applying heat or cold at home to open-back surgery.

Lifting tasks

There are many different types of MMH tasks, such as lifting, holding, carrying, pushing, and pulling. Nearly 50% of such injuries occur while lifting objects, while only 9% occur while holding, wielding, throwing, or carrying objects (Klein, Roger, Jensen, and Sanderson, 1994). Therefore, lifting is considered the most common MMH task associated with the occurrence of low back injuries. This is why the most research studies have focused on the mechanisms and aspects of manual lifting task.

Generally, the goal of those studies has been to develop knowledge in the movement of human body during the lifting trials, and to prevent the risk of injuries, especially in the lower back. Studies have shown that stresses in the lower back during manual lifting to be affected by the speed of the lift and the external loads (Buseck et al., 1987), and also by the techniques of lifting (Busch-Joseph et al., 1988). Those studies concluded that increased speed means increased acceleration, which means increased muscle forces and increased compressive forces on the lower back.

Techniques of lifting

The two extremes of manual lifting technique that are frequently discussed in the literature are squat (leg lifting) and stoop (back lifting). The squat technique is the style where the knees are flexed and the trunk is straight. The stoop is the style where the knees are straight and the trunk is flexed. In reality these techniques are rarely used in their purest form, but rather modifications occur depending on the circumstances of the lift: a freely chosen lifting technique is often called "free-style".

Burgess-Limerick and Abernethy (1997) attempted to quantify the lifting technique using a postural index. This was defined as the ratio of knee flexion from normal standing to the sum of ankle, hip, and lumbar vertebral flexion. The results showed the stooped postures adopted at the start of a lift correspond to postural indices close to 0.0, whereas full squat postures correspond to postural indices close to 1.0. The typical values of postural index for normal (using "free-style") lifting were between 0.57 to 0.61. These postural index results showed that lifting posture was independent of the task characteristics and specific joint positions (load mass and initial load height).

The physiological cost of the three different techniques of lifting (squat, stoop and free-style) in the symmetric and asymmetric planes has been studied (Kumar, 1984). The squat has been shown to be the more physiologically demanding technique in terms of oxygen consumption and inspiratory ventilation volume. The stoop was the least physiologically demanding. The squat technique generated lower biomechanical stresses on lower back than stoop technique (Andersson and Chaffin, 1986). However, this result only works in one condition that the load being lifted must fit between the knees while using the squat technique. If the load is very large that must be lifted in front of the knees, the horizontal distance from the load to the lower back (L5/S1 disc) becomes excessive and greatly increases the stresses on the lower back. Otherwise, using the stoop technique will minimize the horizontal distance, and, will decrease the stresses on lower back.

The efficiency and effectiveness of stoop and squat techniques at different frequencies of lifting have been investigated (Welbergen et al., 1991). The results showed the mean power output and oxygen uptake were greater for squat than stoop techniques at the same frequencies. The mechanical efficiency was defined as the power output divided by the energy equivalence of oxygen uptake. The result was no significant different between the two techniques. The effectiveness was defined as the productive external power output divided by the equivalence of oxygen uptake. And, the result was significantly higher for the stoop technique than squat.

Based on the lifting technique, therefore, different muscle groups will be used to provide necessary net joint forces and moments to perform the lift, and their activities will be different. For example, when using the squat technique, the strength of quadriceps muscles will be a limiting factor. This is because of the muscle cross sectional area of the erector spinae muscle is larger than that of the quadriceps muscles. So, it can be anticipated that at heavier loads the stoop technique would be adapted (Davis et al., 1965). The changes in lifting technique as a function of the amount of weight lifted also has been discovered (Schipplein et al., 1990). The results showed that the lifting technique changed from more of a squat to more of a stoop as a result of biomechanical stresses when the weight of external load was increased. It was also discovered that lifting technique changes from more of a squat lift to more of a stoop lift when the quadriceps muscles were fatigue (Trafimow et al., 1993).

Biomechanical models

There are several methods that could be used to estimate compressive force in the lower back. The different types of biomechanical analyses of stresses on the lower back during manual lifting include direct measurement method, simulation models, finite element method, electromyography (EMG), top down and bottom up approach.

Intradiscal pressure measurement is the most direct and reliable method for assessing loads on the spinal motion segment. Nachemson and Elfstrom (1970) first described this measurement system, which consisted of a pressure transducer attached to the tip of a needle inserted between vertebral discs. Disc-pressure measurements also have been used to validate biomechanical models from which the loads on the lumbar spine were predicted (Schultz et al., 1982). The results of this experiment, the predicted load and measured quantities correlated well. It was shown that such invasive measurements could be avoided and replaced by prediction from a biomechanical model under a variety of conditions.

A stochastic (probabilistic) model of trunk muscle activation was developed to quantitatively capture the trunk muscle variability during bending motions, such as those involved in lifting (Mirka and Marras, 1993). The model was based on a simulation of experimentally derived data and predicted the possible combinations of time-dependent trunk muscle coactivations that could be expected given a set of trunk bending conditions. The results indicated that the variability in trunk muscle force had a small effect on spinal compression variability (+/- 7% of mean compression), but greatly influenced both lateral (+/- 90% of mean) and anteroposterior shear forces (+/- 40% of mean).

A biomechanical simulation model has been developed to obtain a tool for analyzing the relations between forces in muscles, ligaments and joints in the transfer of gravitational and external load from the upper body via the sacroiliac joints to the legs in normal situations (van Dijke et al., 1999). The model comprises 94 muscle parts, 6 ligaments and 6 joints. The model enables the calculation of forces in the pelvic structure for different postures based on linear/non-linear optimization scheme.

The finite element method treats the lumbar spine disc as a geometric mesh of tensile elements and assigning material properties according to the type of biological structure being defined.

As the exterior loading inputs are altered, the mesh will deform as a function of the geometry and the material stiffness properties assigned to each element. The deformation in the mesh is compared with experimental results using discs from cadavers under similar loading conditions. These models can be valuable in predicting which components of the lumbar spine are the most highly stressed and are at risk of being injured (Lavaste et al., 1992, Rao and Dumas, 1991, Shirazi-adl, 1989).

EMG activity of the erector spinae (back and fascia) muscles, using corrections for muscle length, contraction velocity and electro-mechanical delay were used to calculate the extensor moment acting on the lumbar spine during lifting activities (Dolan and Adams, 1994). The results of this study showed that the techniques of lifting, mass of the load, horizontal distance in front of the feet, and speed of movement influenced the peak extensor moment on the lumbar spine. The stoop lifting generated 10% lower of the peak extensor moment compared to squat lifting. Extensor moment increased substantially with increasing mass, horizontal distance from the feet, and the speed of movement.

The various types of biomechanical models assume the human body to be composed of dynamic, rigid, and linked segments. These types of models apply laws of motion and mechanical properties to estimate stresses incurred in the body while lifting. Many types of these models focus primarily on the lower back, specifically the lumbosacral vertebrae, because of the high incidence of injuries in that area of the body. Based on statistics, it was shown that between 85% and 95% of all disc herniations occur which relatively equal frequencies at the L4/L5 and L5/S1 levels (Smith et al., 1944, Armstrong, 1965, Krusen et al., 1965).

In order to study these types of biomechanical models of manual lifting, especially the stresses that occurred in the lower back, the kinematics, kinetics and anthropometry data information are required:

Kinematics

The description of the human movement, independent of forces that cause the movement is called kinematics. They include linear and angular displacements, velocities, and accelerations. The displacement data are taken from any anatomical landmark: center of gravity of body segments, centers of rotation of joints, extremes of limb segments, or key anatomical prominances. The velocities and accelerations data are derived from the displacement data.

Investigators in human movements, especially in lifting motion were interested usually to use video-based systems to track the positions of a variety of markers placed on the body, as follows:
A bolex sixteen-millimeter camera and black theatrical greasepaint markers, which were manually digitized to detect the position of markers during lifting motions (Patterson, et al., 1987).
Stereocinematography with two locam sixteen-millimeter cameras to analyze the position of anatomical markers during asymmetrical lifting (Gagnon and Gagnon, 1992).

Kinetics

The study of the forces and the resultant energy generated by the body during human movement is called kinetics. Knowledge of the patterns of forces is necessary for an understanding of the cause of any movement. The process by which the reaction forces and muscle moments are calculated is called link-segment modeling. This link-segment model works based on some assumptions, as follows:

1. Each segment has a fixed equivalent mass located as a point mass at its center of mass (which will be the center of gravity in the vertical direction).
2. The location of each segment's center of mass remains constant within the segment during the movement.
3. The joints are considered to be frictionless hinge (or ball and socket) joints.

4. The mass moment of inertia of each segment about its mass center (or about either proximal or distal joints) is constant during the movement.

5. The length of each segment remains constant during the movement.

In the link-segment model, the human body is assumed to be a chain of rigid segments. These segments are interconnected by joints, which are considered as hinge joints. Conventional Newtonian mechanics are applied to each individual segment, starting at the distal or proximal end of the chain, and intersegmental reactive moments and forces are calculated, considering inertial as well as gravitational effects.

In order to measure forces exerted by the body on an external body or load, a suitable force-measuring device is required. A device, called a force transducer, gives an electrical signal proportional to the applied force. There are many kinds of force transducers available: strain gauge, piezoelectric, piezoresistive, capacitive, and others. The most common force acting on the body is the ground reaction force, which acts on the foot during the motion, such as lifting, walking, and running. A force platform constructed with strain gauges was used to analyze of forces and torques on the lower back joint between static and dynamic models of lifting tasks that performed in four different lifting techniques (Leskinen, 1985). The piezoelectric platform has been used to study the relationship of the joint moment magnitude at L5/S1 disc, hips and knee joints while lifting (Schipplein et al.,1990).

The link-segment biomechanical model can be applied in different starting point of analysis, as follows:

Top-Down approach

The Top-Down approach (Leskinen et al., 1983, Freivalds et al., 1984, Kromodihardjo and Mital, 1986) starts applying first equation to the hands/load segment, assuming no external forces besides gravity acting in this segment. Next, applying the equation to links respectively to forearms, upper arms, and trunk segments to assess moment and force acting at an intervertebral joint. In this analysis no force-platform data are used.

Bottom-Up approach

The Bottom-Up approach (Busch-Joseph et al., 1988, Buseck et al., 1988, Schipplein et al., 1990) starts applying the equations to the feet. Next, the procedure is the same as the top-down approach, but in the opposite direction by applying the equations to links representing the lower legs, upper legs, and pelvis respectively, moment and force at one of the intervertebral joints are calculated. In this approach, force platform data are required to generate ground reaction forces. When the analysis starts at the feet, the sequential application of equations ends at the hands/load segment.

The lifting tasks in this study were done using the bottom-up approach to calculate the stresses on the lower back during manual lifting. A study of validation of two-dimensional dynamic linked segment model either using Top-Down or Bottom-Up approach to calculate joint moment at L5/S1 level in lifting was done (de Looze et al., 1992). The reactive forces and moments were calculated both by started once at the feet (ground reaction forces) and started once at a hands/load segment. The moments at L5/S1, calculated starting from the feet compared to starting from hands/load yielded a coefficient of correlation (r) equal to 0.99.

Anthropometry

In order to increase the accuracy of these types of biomechanical models. Subject anthropometry data were required to take into account the different body shapes and sizes of individuals. The human movement analysis requires kinetics measures as well: masses, moment of inertia, and the center of mass locations. Some previous studies of developing the anthropometric data measurement (segment inertial parameter) are as follows:

Eight unfrozen cadavers were used to investigate the center of mass, weight, and moment of inertia of the body segments (Dempster, 1955). Clauser et al. (1969), then did similar segment characteristics to Dempster, but used thirteen frozen cadavers. Finally, Chandler et al. (1975) studied on same segment parameters from six cadavers, plus, found the joint centers of rotation by dissection through segmentation planes.

Zatsiorky and Seluyanow (1983) determined the center of mass, weight, and radii of gyration for body segments from 100 live subjects using a gamma-ray scanning technique. This segment parameter model was not used for the biomechanical model in this research. This was because this model used bony landmarks as reference points for locating segment center of mass (CM) and defining segment lengths. Some of these points are remarkably distant from the centers of the neighboring joints. As consequence, when subject flexes his joints the distance of these reference points from the respective proximal or distal segment CMs significantly decrease. These and other related changes, which make it impossible to accurately locate segment CMs. de Leva (1996) revised the Zatsiorsky and Seluyanow (1983) segment inertia parameters to reference joint centers rather than bony landmarks.

The biomechanical model of the current study was used de Leva (1996) segment inertia parameter's values, except for the pelvis (lower trunk) segment. This was because de Leva defined the lower trunk segment as the segment between the omphalion and the hip joint center. The model of current study utilizes a pelvis segment defined by the hip and L5/S1 joints.

Static and dynamics models

In the early twentieth century, most of the researches of manual lifting were static, thus revealing the postural stress due to gravity. When using static models the effects of acceleration are assumed negligible which leads to a large underestimation of the spinal stress in dynamic activities, such as lifting. In dynamic models, the inertial forces and torques induced by acceleration, based on the initial and final positions and the total displacement time of each joint are involved. Some studies have compared static and dynamic models in lifting, as follows:

The peak compression force on the lumbosacral joint of dynamic model for four different lifting techniques were 33% to 60% higher than in the static model depending on lifting technique (Leskinen, 1985). Tsuang, et al., (1992) showed the flexion-extension moments at L5/S1 level and hip joint using static and dynamic analysis for two different weights (50 N and 150 N) and two different speeds (normal and fast). The results showed the difference between dynamic and static analysis was greatest when lifting 50 N at a fast speed: an 87% increases in L5/S1 moment and 95% increases in hip joint moment was observed when replacing the static with a dynamic analysis.

Two-dimensional and three-dimensional dynamic models

The two-dimensional dynamic lifting model (Busch-Joseph et al., 1988, Buseck et al., 1988, de Looze et al., 1992, Schipplein et al., 1990) restricts analysis to the sagittal plane only, and assumes that the movement is symmetrical, considering only two-dimensional movement flexion and extension of the back. The movement that occurs about the longitudinal axis of the segment is negligible.

The three-dimensional dynamic model is a complicated model that includes twisting of the back and asymmetrical motions. A dynamic three-dimensional multi-segment model to compute spinal loading (torsion, flexion/extension, and lateral-bending moments) for asymmetrical lifting and lowering tasks performed in different speeds has been developed (Gagnon and Gagnon, 1992). The results showed that lifting tasks, in the fast and accelerated conditions, generated significant increases over the slow condition for torsion, flexion/extension, and lateral-bending moments in the L5/S1 disc. A three-dimensional and dynamic model of the human body for lifting motion, which has been developed to predict the T10/T11 intervertebral joint moment (Gillette, 1999). This study investigated a three-dimensional model to predict joint forces and moments at T10/T11 as an upper body and a lower body formulation. The model of this study was also used to compare between symmetric and asymmetric lifting for both leg-lift and back lift techniques.

Muscular fatigue and electromyography (EMG)

Fatigue, generally defined as a transient loss of work capacity resulting from preceding work, produces a general feeling of discomfort and frustration and interferes with well being and performance. Muscular fatigue specifically is defined as a reduction in maximal production of the muscle and is characterized by a reduced ability to perform work.

Fatigue can be classified as central or peripheral on the basis of the location of the site of fatigue. Possible causes for central fatigue include a malfunction of the nerve cells or inhibition of voluntary effort in the central nervous system. Other factors that may influence central fatigue include psychological factors and motivation that influence how much effort an individual will give, particularly if pain is associated with continuing an activity. Peripheral fatigue refers to fatigue at a site beyond the central nervous system; this may include sites within the peripheral nervous system or within the skeletal muscle. Peripheral fatigue can occur at several sites: the neuromuscular junction, the sarcolemma-T-tubules-sarcoplasmic reticulum system. It is also possible that fatigue results from biochemical and metabolic changes within the contractile elements (myofilaments of the muscle cell).

Electromyography (EMG) is defined as the recording of action potentials emitted from contracting muscles (Chaffin and Andersson, 1999). There are two types of electrodes of EMG, within the muscle (i.e., an indwelling or intramuscular electrode), and on the skin (i.e., surface electrode). The relationship of EMG activity to muscle force, which depends on several factors, appears to be monotonic, in the sense that an increase in tension is paralleled by an increase in myoelectric activity. However, this relationship is nonlinear under many circumstances.

EMG signals processed by means of so-called full-wave rectification have been used to study EMG force relationships. These relationships are often empirically described with piecewise linear regressions (Chaffin et al., 1980) or by using power functions of the form $IEMG = a(F)^b$, where the parameters a and b are fit by least squared error regressions (Kumar, 1996).

The EMG is used to evaluate not only relative muscle activity, but also localized muscle fatigue. Such an estimate of muscle fatigue relies on changes in the spectral and amplitude. characteristics of the EMG. A comparison of the electrical activity from the biceps brachii during three stages of a fatiguing static contraction has been studied. The results showed both increase in amplitude and decrease in median frequency with fatigue (Zuniga and Simon, 1969).

The effect of fatigue conditions

Lifting objects is certainly a physically exerting task, especially in the lifting tasks that must be done repetitively. Repetitive lifting has been shown to be a risk factor for the development of low back injury (Frymoyer et al., 1983). Some experimental studies have shown that repetitive lifting causes a rapid development of back extensor muscle fatigue (Petrofsky and Lind, 1978; Potvin and Norman, 1993). Muscle fatigue has also been defined as any reversible decrease in the performance capacity of a muscle that results from its activity (Bigland-Ritchie and Woods, 1984).

The particularly high risks associated with repetitive lifting may be cause of excessive lumbar flexion with resulting fatigue on the back muscles investigated by Dolan and Adams (1998). The erector spinae muscles protect the spine from excessive flexion, but in so doing they impose a high compressive force on it. If these muscles become fatigued, and therefore less able to generate high forces quickly, then the moment (bending) acting on the lumbar spine may increase, and the compressive force decrease. Fatigue in the erector spinae muscles was quantified by comparing the frequency content of the EMG signal during static contractions performed before, and immediately after repetitive lifts.

Another investigation of the repetitive lifting tasks showed fatigue that generated from repetitive lifting tasks was documented as the reduction in the average lifting force of hip joint and spine torque generation. Fatigue was also associated with decreased knee and hip motion. and increased lumbar flexion (Sparto et al.. 1997). The fatigue condition was defined either as subjective opinion (subject thought that he could no longer continue to lift) or when the heart rate of the subject attained 180 beats per minute.

Chen (2000) found that the peak compressive force on the L5/S1 disc was increased when localized arm fatigue was generated. The fatigue condition of this study was generated by requiring the subject to stand erect with shoulders flexed at ninety degrees and elbows fully extended while holding a 5-kg barbell with both hands until the subject was unable to maintain the posture any longer (between 25 and 40 seconds).

Trafimow. et al. (1993) studied the effect of quadriceps fatigue on the technique of lifting. The fatigue protocol was applied by asking the subjects to stand in a half squat until they were unable to hold the position (between 60 and 90 seconds). Weights from 0 N to 300 N were lifted in 50-N increments. Subjects tended to shift from a squat technique (most angular motion occurring in the leg and thigh segments) towards a stoop (most angular motion occurring in the pelvis and trunk segments). In spite of these kinematic changes there was no significant difference in the L5/S1 moment pre and post fatigue.

The effect of previous lifts and the knowledge of load magnitude

Occupational lower back pain has emerged as a serious clinical. social and economic problem in industrial world (Pope et al.. 1991). In a large proportion of patients. it is currently unclear what causes the pain. Efforts to reduce the incidence of low back pain in the workplace are often based upon the evaluation of manual materials handling tasks. in which several load determining factors are involved. e.g. the load location. the displacement of the load. the asymmetry of lifting. the lifting frequency. and the coupling between load and hands. The NIOSH equation provides a method for computing a weight limit for manual lifting from these factors (Waters et al.. 1993). The absence of load knowledge, however, is a factor that is not accounted in this equation, or in any other evaluation approach. The daily situation of workers like refuse collectors, luggage dispatchers and movers is fundamentally different from the situation in which an unsuitable preparation is achieved by giving false expectations. Epidemiological studies have shown that workers exerting sudden unexpected maximal efforts are particularly vulnerable to low back disorders (Magora, 1973).

Some previous experimental studies have compared conditions where people either knew the actual mass or did not know, are as follow:

Patterson et al. (1987) studied the peak L4/L5 moment during lifting masses of 6.8, 10.2, and 13.8 kg under known and unknown the knowledge of the load. They observed a general tendency towards greater peak moments in the condition of not knowing the actual mass. In this study, there were two different groups of subjects, one group of experienced weight lifters and one group of novice weight lifters.

Butler et al. (1993) estimated the peak L5/S1 moment for lifting loads of 0, 15, 25 and 30 kg. Only at 0 kg was the peak moment 30% greater in the condition of not knowing the actual mass as compared with the known condition. At the other masses, no significant difference was found.

Commissaris and Toussaint (1997) studied the moment at L5/S1 with a sudden change of lighter mass (6 kg) after a long series lifts of heavier mass (16 kg). The moment at L5/S1 was not different between lifting the 6 kg and the 16 kg, until 150 ms after the box lift-off. In this study, subjects were not informed about the sudden changes of load mass that were going to take place. So, subjects were induced to overestimate the mass to be lifted.

de Looze et al. (2000) studied the lower back moments between the condition of knowing actual mass (known condition) and the condition of knowing only the range of masses to be lifted (unknown condition). The peak L5/S1 moments in the unknown condition were greater (10%) than in the known condition. The range of masses in this study was between 6.5 to 16.5 kg.

der Burg and Dieen (2001) showed the L5/S1 moments and compression forces were not different when subjects were lifting an unexpectedly heavier object (underestimation). In this study, subjects performed series of 10 lifts of lighter mass (1.6 kg or 6.6 kg), and at the end of lifts a heavier mass (11.6 kg or 16.6 kg) was unexpectedly placed. So, subjects were induced to underestimate the load to be lifted.

The effect of constraining barrier

Many manual material handling tasks require individuals to lift from, and lower into, industrial bins. These bins or barriers can serve to constrain the flexion of the knee and this can have implications for spine flexion during load handling. However, only one study looked at the effects of constraining barrier on lifting and lowering, which has implications for many material handling tasks found in industry (McKean and Potvin, 2001). In this study, the posture and muscle activity (EMG) of the lower back during lifting and lowering loads either placed behind a constrained (120% of knee height) or freestyle (no barrier) conditions were investigated. The main finding of this study was that the constraint barrier appeared to increase the risk of injury during lifting and lowering by increasing horizontal reach, trunk flexion and magnitude of muscle activity loading.

CHAPTER 3: INFLUENCE OF FATIGUE ON L5/S1 COMPRESSIVE FORCE DURING MANUAL LIFTING

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Abstract

The purpose of this study was to examine the mechanical stresses on the lower back of subjects performing a lifting task until fatigued. It was hypothesized that subjects would have individual strategies for dealing with fatigue. Overall, results showed that the peak compressive forces on the lower back during the lifting phase were not significantly different between pre and post fatigue. Variables that were significantly different during the time of peak compressive force were the vertical acceleration of the load, the horizontal distance between the center of the load and the L5/S1 joint, the angular acceleration of the trunk, and the height of the load. During fatigue, 5 of the 13 subjects increased the peak compressive forces. The reasons were due to increase in the vertical acceleration of the load and the angular acceleration of the trunk, and shift toward a stooped posture when fatigued. But, 8 of the 13 subjects decreased the peak compressive forces when fatigued. The subjects in this group had smaller increases in the vertical acceleration of the load and the angular acceleration of the trunk. They also decreased the horizontal displacement of the load relative to the L5/S1 joint, increased the sacral angle, and maintained posture when fatigued.

Relevance to industry

Workplace activities rely on the lifting of loads, often until a fatigue condition is reached. This study investigated the strategy-adopted response to changes in fatigue condition and their effects on the

L5/S1 compressive forces during lifting. The results have implications for lifting job design and provide useful information for further study in the prevention of low back injuries.

Keywords: Lifting, fatigue, L5/S1 compressive forces.

Introduction

Biomechanical models have often been used to estimate spinal compressive forces during lifting activities (Chen, 2000, de Looze, et al., 1992, Gillette, 1999, Schipplein et al., 1990, Sparto et al., 1997, Trafimow et al., 1993). These estimations are based on knowledge of the external forces acting on the body as well as anthropometric and kinematic data. These data are combined in a rigid-body, linked-segment model in which the equations of the motion are used to estimate joint reaction forces and moments. When combined with knowledge of lower back musculature and skeletal alignment the compressive and shear components of force in the lower back can be estimated.

Epidemiological studies have identified manual lifting as a major factor in the occurrence of low back injury (Chaffin and Park, 1973, Troup 1965). It is generally accepted that this is due the mechanical effects on the lower back, so appropriate predictions of the lower back stresses during lifting are an important prerequisite towards addressing problems of lifting induced back injury. Repeated lifting activities increase the risk of developing low back injuries (Kelsey et al., 1984, Magora et al., 1973, Marras et al., 1993). Lifting develops a high compressive force on the lumbar spine (Dolan et al., 1994), which can cause lumbar discs to prolapse posteriorly, especially if the forces are applied repetitively (Adams and Hutton, 1985). The National Institute for Occupational Safety and Health (NIOSH, 1981) has recommended that predicted L5/S1 compression force values above 3400 N be considered potentially hazardous for some workers. If the values are greater than 6400 N, the job is hazardous to most workers.

Fatigue can influence lifting kinematics and the peak moment and compressive forces at L5/S1 (the joint between the fifth lumbar and the first sacral vertebrae) during manual lifting.

Trafimow, et al. (1993) studied the effect of quadriceps fatigue on the technique of lifting. The fatigue protocol was applied by asking the subjects to stand in a half squat until they were unable to hold the position (between 60 and 90 seconds). Weights from 0 N to 300 N were lifted in 50-N increments. Each weight was lifted once from the floor to knuckle height before and after the fatigue protocol. Subjects tended to shift from a squat technique (most angular motion occurring in the leg and thigh segments) towards a stoop (most angular motion occurring in the pelvic and trunk segments). They also reduced peak knee and hip flexion and increased peak trunk angular velocity. In spite of these kinematic changes there was no significant difference in the L5/S1 moment pre and post fatigue.

The effects of arm fatigue have also been investigated (Chen, 2000). Fatigue was induced by requiring the subjects to stand erect with shoulders flexed at ninety degrees and elbows fully extended while holding a 5-kg barbell with both hands until the subjects were unable to maintain the posture (between 25 and 40 seconds). The subjects lifted weights ranging from 5 to 30 kg in 5 increments. Each weight was lifted once either from floor to knuckle height or from floor to shoulder height, both before and after the fatigue protocol. The peak compressive forces at L5/S1 were increased after the localized arm fatigue was generated. Also, when subjects lifted less than 20 kg the peak acceleration of the load increased with fatigue. The subjects also shifted from a squat lift toward a stoop lift when the arms were fatigued.

Sparto, et al. (1997) documented a reduction in knee and hip range of motion and an increasing in the spine peak flexion during a lifting test. The task was performed using a lifting device to simulate the inertial components of a box with weight equal to 25% of the subject's maximal lifting capability from mid-shank to waist level. Subjects continuously lifted until they felt they could no longer continue, or until their heart rate exceeded 180 beats/min. Fatigue was documented as a reduction in average lifting force and hip and spine moment generation.

Fatigue has also been studied when the fatiguing protocol was actual lifts (Dolan and Adams, 1998). Subjects lifted a 10 kg load from floor to waist height 100 times. These researchers used erector spinae electromyography (EMG) in the estimation of compressive forces. Results showed that peak lumbar flexion increased during the 100 lifts, but the peak spinal compression decreased.

These four studies highlight some of the difficulties associated with back research during fatigue. First, there are many ways of fatiguing the body. On a continuum between complete experimental control and complete generalizability of results, the first two studies chose a fatigue protocol that emphasized control. Particular parts of the body were systematically fatigued so that subjects with different lifting styles would not preferentially fatigue different parts of the body. The third and fourth studies used multiple lifts to fatigue the subjects. In this case, the part of the body that was fatigued was largely unknown but the dynamic lifting protocol was more consistent with actual workplace lifting conditions. In addition, the first three studies used a protocol that fatigued the subjects until volitional failure, the subjects in the fourth study lifted 100 times and therefore the magnitude of fatigue probably differed between subjects.

In general, back research has focused on group designs in which the results of the entire sample are pooled. This assumes that all subjects are behaving in a consistent manner. There is no ability to assess individual strategies that may cause some subjects perform in a manner opposite of other subjects. In the group design this would cause a washout effect in which a performance in one direction is offset by a performance in the opposite direction. The net result is that no significant differences are found.

The purpose of this study was to examine the mechanical stresses on the lower back of subjects performing repetitive lifting tasks until fatigued. Group differences were examined to determine generalized results and individual results were looked at to identify strategies adopted by individuals or groups of individuals. It was expected that the subjects would differ in many important ways such as anthropometrics, strength, endurance, lifting style and motivation. It was hypothesized that these characteristics would lead to different individual strategies to deal with fatigue.

Methods

Subjects

Thirteen healthy young male subjects with no history of back injuries participated in this study. The mean age was 22 ± 1.6 years; body mass was 77 ± 12.9 kg; and body height was 1.78 ± 0.09 m. The subjects signed informed consent to participate in this study in accordance with university policy. Prior to the start of the study, subjects were familiarized with the experiment protocol.

Protocol

Each subject was asked to lift a crate (37.5 cm length, 33 cm width, 27 cm height) containing a mass of 10 kg from the floor to knuckle height directly in front of the subject. The crate had two fixed handles placed symmetrically 27 cm above the bottom. The handles and mass center of the crate were positioned approximately 27 cm horizontally from the subject's ankles. Each subject wore shoes during the trials. The right foot of each subject was placed on the force platform during the trial experiment.

Subjects began the lifting task from the standing position. From the standing position, then bent down, grabbed the crate and returned to a standing position as they lifted the crate to a knuckle height. The crate was then returned to the initial position. Each subject was asked to lift the crate at a 'normal' speed using a freestyle technique, i.e. the technique that was the most comfortable for each subject. A lift was initiated every 6 seconds based on an audible tone.

A pre-fatigue condition consisted of 10 successive lifts during the first 2 minutes of lifts. Every minute the subject rated the comfort level on a rating of perceived exertion (RPE) with a scale from 0 (no discomfort) to 10 (extremely uncomfortable). A 10 on this scale would indicate that the subjects would be unable to continue lifting. After the subjects reached at 9 on the scale, they were asked to perform another ten lifting trials while data were recorded. The final 10 lifts comprised the post-fatigue condition.

Model

Reflective Markers were placed on the second toe, the posterior point of the heel, the lateral malleolus, the midpoint of the lateral joint line of the knee, and the glenohumeral joint. The three additional markers were placed on the pelvis to help define the joint between the fifth lumbar and first sacral vertebrae (L5/S1). These markers were located at the end of an 8-cm wand placed over the anterior iliac crest and over the posterior sacrum. These markers were used to form a local coordinate system with the hip marker as the origin. A final marker was placed on the crate.

Three-dimensional kinematic markers for this experiment were digitized using a Peak Video and Analog Motion Measurement System (Peak Performance Technologies, Inc., Englewood, CO). Four 120 Hz video cameras were used to obtain the positions of the reflective markers. The marker coordinates were low-pass filtered with a fourth-order (zero lag) Butterwoth filter using a 2 Hz cut-off frequency.

Before performing the lifts, each subject was asked to stand in a static erect position while kinematic data were recorded. The L5/S1 sagittal position was estimated to be 19.5% from the hip to the shoulder (Lanier, 1939 and de Looze, 1992). The sacral angle (α) was defined as the angle formed by the base of sacrum (Chaffin, et al., 1999). It was assumed to be 45° during static trial (Thieme, 1950). The L5/S1 coordinates were then transformed into a local pelvis coordinate system with the hip as the origin. The location of L5/S1 and α were assumed to be constant in the local coordinate system during the lifting task.

A 5-segment rigid body model was developed using the foot, leg, thigh, pelvis, and head/arms/trunk (above L5/S1). It was assumed that these segments were connected by frictionless pin joints, namely the ankle, knee, hip, and L5/S1 joints. Each segment was assumed to have a fixed point mass. It was also assumed that the moment of inertia about the center of mass was constant. Center of mass locations within the segments and the moments of inertia were approximated using anthropometric data and equations obtained from de Leva (1996). The only exception was the pelvis segment, which de Leva defined the lower trunk segment between the omphalion and the hip joint center. The current model utilized a pelvic segment defined by the hip and L5/S1 joints. Therefore the mass of the pelvis segment was estimated to be 11.8% of body mass (Web Associates, 1978). Center of mass and moment of inertia values were modeled as an elliptical solid (Hanavan, 1964) with dimensions determined from anthropometric measurements. The sensitivity of peak compressive force values to errors in the pelvis center of mass and moment of inertia was estimated by increasing these values by 10% during the pre-fatigue condition of one of the subjects. This increased the peak compressive force by less than 0.5%.

A strain-gage force platform (Advanced Mechanical Technology, Inc., Newton, MA model OR 6-6 2000) was used to determine the location, direction and magnitude of external ground reaction forces. The force platform signals were sampled at 120 Hz and synchronized with the kinematic data using the event and video control unit (Peak Performance Technologies, Inc., Englewood, CO).

The reaction forces and moments at each joint were calculated using inverse dynamics for the standard link segment model (Winter, 1990). Newtonian equations of motion were applied to each individual segment starting at the foot. Reaction forces and joint moments were then estimated for the proximal end of each segment and the end of process was then repeated for the next segment in the model. Equations of motion and the anthropometric model were implemented using a custom analysis program.

To calculate the compressive force on the L5/S1 joint, the following simplifications were used (Chaffin, et al., 1999):

1. No abdominal pressure acted on the diaphragm in front of spinal column (Chaffin, et al., 1999, de Looze et al., 1999).
2. The line of action of the extensor spinae muscles of the lower back was assumed to act parallel to the normal force of compression on the L5/S1 joint and with a moment arm (E) of 6.0 cm (Kumar, 1988).
3. The compression force was assumed to act at the center of rotation of the joint and thus was not considered in the moment equation.

Muscle force (F_M) was then solved for as follows:

$$F_M = M_{L5/S1} / E$$

Finally, the forces acting parallel to the disc compression force (F_C) were expressed by

$$F_C = F_M + F_V \text{ cosine}(\alpha) + F_{AP} \text{ cosine}(\alpha)$$

Where, α was the sacral angle, F_V was the vertical reaction force, and F_{AP} was the anteriopostior reaction force.

Electromyography (EMG)

To quantify the muscular fatigue experienced by the subjects, pairs of skin surface electrodes (Mediatrace) were attached longitudinally over the subject's lower back. The interelectrode distance was approximately 4 cm. The EMG signal was recorded at 1000 Hz from parts of the erector spinae muscles at electrode locations according to Roy et al. (1989) (3-cm lateral to the spinous process at L5). The ground electrode was attached to the left side of distal end of ulnar styloid process. The EMG signals were collected for 10 lifts of pre and 10 lifts of post fatigue conditions. The analog

EMG signals generated from the lower back muscle activities during the lifts were preamplified (100 x) and high-pass filtered at 10 Hz to remove movement artifact. The EMG signals were digitized by a portable data logger (Model BM42. Biomedical Monitoring. Glasgow. and U.K.). The EMG was digitally low-pass filtered at 490 Hz and burst activity from each lift was identified. Since the EMG bursts contained a complete cycle of data. nonstationarities within a burst could be disregarded (Bonato. et al.. 2001). A fast Fourier transform was used to examine the EMG signal in the frequency domain. The median frequency (van Dieen. et al.. 1993) and the root mean square (RMS) amplitude (de Luca. 1997) were calculated for each pre fatigue and each post fatigue lift. It has been shown that decreases in median frequency and increases in RMS amplitude are indicative of muscular fatigue (van Dieen. et al.. 1993 and Zuniga and Simon. 1969).

Data analysis

The time of the peak compressive forces at L5/S1 was observed slightly after the instant that subjects lift-off the load. An example of the peak compressive force at L5/S1 from one subject in the pre fatigue condition is shown in figure 1. There are 10 peak pairs that are relatively close to each other. The first peak of each pair represents the lifting of the load and the second peak represents the lowering. In between each pair of peaks is a period of time in which the subject is in a standing position and waiting for the next lift. In order to analyze the differences between pre and post fatigue lifting several variables were examined during the time of peak compressive loading. These included vertical load acceleration. trunk angular acceleration. horizontal distance between the L5/S1 joint and the center of the load. height of the load and sacral angle and the posture index (PI). The postural index was defined as the ratio of the knee angle to the sum of the hip and L5/S1 angles. Burgess-Limerick and Abernethy (1997) used a similar index to identify lifts on a continuum from complete stoop lifting (PI = 0.0) to complete squat lifting (PI= 1.0).

Means and standard deviations were calculated for pre fatigue and post fatigue conditions. Repeated measures analysis of variance was performed to detect statistically significant differences between the conditions. Subjects were then grouped according to whether peak compressive forces increased or decreased with fatigue. Additional means and standard deviations were calculated for these groups.

Results

Overall discomfort during the lifts was documented using an RPE scale from 0 to 10. On average, subjects lifted for 24.4 ± 18.5 minutes. This time was one minute after they reached a 9 on the RPE scale. Two of the 13 subjects did not reach an RPE score of 9 prior to the 1-hour time limit. Both of these subjects reached an 8 on the RPE score and both were left in the analysis. The magnitude of muscular fatigue was documented by changes in the median frequency and RMS of the electromyography. During pre fatigue, the mean median frequency of the EMG was 69.2 ± 4.7 Hz. This decreased to 59.9 ± 4.9 Hz after fatigue ($p = 0.003$). The mean RMS of the EMG was 0.10 ± 0.01 mV in pre fatigue and the value increased to 0.12 ± 0.02 mV after fatigue ($p = 0.019$). Three of the subjects did not have EMG data due data collection errors.

Peak compressive forces at L5/S1 were observed slightly after the load left the ground (Figure 2). Analysis of variance did not show a statistically significant difference between pre (3821.9 ± 484.9 N) and post (3769.6 ± 468.4 N) fatigue conditions ($p = 0.543$).

There were some statistically significant ($p < 0.05$) kinematic changes that occurred during fatigue that were relatively consistent across subjects (Table 1). These variables were all measured at the time of peak L5/S1 compression. Vertical load acceleration increased in 12 of the 13 subjects and trunk angular acceleration increased in 11 of the 13 subjects when they became fatigued. On average, vertical load acceleration increased by 28% (2.5 ± 2.2 m•s^{-2} vs 3.2 ± 2.9 m•s^{-2}) and trunk angular acceleration increased by 51% (138.7 ± 74.3 deg•s^{-2} vs 209.6 ± 115.4 deg•s^{-2}). In addition, horizontal displacement of the load increased in 12 of the 13 subjects and vertical displacement of the load increased in 7 of the 13 subjects when they became fatigued. Relative to the L5 S1 joint, the load trajectory was 3 cm closer horizontally (0.57 ± 0.04 m vs 0.54 ± 0.04 m) and 2 cm higher (0.35 ± 0.04 m vs 0.37 ± 0.07 m) during the fatigue condition. The sacral angle and the postural index did not show statistically significant changes with fatigue.

There was no statistically significant difference in peak compressive force; however, five of the subjects showed a 7.4% increase in peak compressive forces (3664.3 ± 195.3 N vs 3935.3 ± 271.5 N) (Table 2) while the other eight subjects showed a 6.5% decrease (3920.4 ± 593.6 N vs 3666.0 ± 549.5 N) (Table 3). There was little difference between these two groups in mass (78.8 ± 13.6 kg vs 76.0 ± 13.4 kg) or height (1.8 ± 0.1 vs 1.8 ± 0.1 m). However, the group that increased peak compressive forces lifted, an average, 35.0 ± 24.1 minutes while the group that decreased peak compressive forces lifted, an average, 17.8 ± 11.2 minutes. Subjects that increased peak compressive forces tended to have greater increases in vertical load acceleration (54.2%) than did subjects having decreased peak compressive forces (16.0%). They also maintained their sacral angle during fatigue (63.0 ± 17.6 degrees vs 62.4 ± 15.4 degrees) while the subjects that decreased peak compressive forces tended to increase their sacral angles (50.6 ± 12.7 degrees vs 55.0 ± 10.4 degrees). Another difference between these groups was the postural index. Subjects that increased peak compressive forces decreased the postural index from 0.50 ± 0.22 during pre fatigue to 0.45 ± 0.21 during post fatigue, indicating a lifting style that is shifting from squat to stoop on the continuum. Subjects that decreased peak compressive forces during fatigue showed a much smaller shift (0.65 ± 0.14 in pre fatigue vs 0.64 ± 0.14 in post fatigue).

Discussion

This study used a dynamic biomechanical model of lifting to predict changes in kinematic and kinetic data obtained under conditions of pre and post fatigue. The purpose was to examine the mechanical stresses (compressive forces) on the lower back of subjects performing repetitive lifting tasks until fatigued. It was hypothesized that subjects would have individual strategies to deal with fatigue. The results showed that the individual lifting strategies changed during fatigue and resulted in increased peak compressive forces in some subjects and decreased peak compressive forces in others.

Repeated lifts of a 10 kg load produced fatigue in most subjects within 1hour. In 11 of the 13 subjects, the discomfort associated with repeatedly lifting the load was of a great enough magnitude to be very near to causing a voluntary cessation of lifting (9 out of 10 on a discomfort scale). The two subjects that did not reach this level of discomfort did reach 8 out of 10 and were therefore considered adequately fatigued to include in the study. Most of the subjects (11 out of 13) indicated that back discomfort was the limiting factor in the duration of the lifting session. One possible source of the discomfort was fatigue in the lower back muscles. Previous studies have shown that median frequency decreases and the RMS amplitude of the EMG increases (van Dieen, et al., 1993, Zuniga and Simon, 1969), when muscles become fatigued. The EMG results in this study are consistent with muscular fatigue in the muscle group that extends the back at the level of L5/S1. There are problems with this type of analysis. There is no guarantee that the motion that is occurring during the pre-fatigue is the same as the motion during the post fatigue. In. fact we have demonstrated that there are kinematic and kinetic changes that occurring during fatigue. It is difficult to attribute the changes in EMG, especially the amplitude changes, to local muscular fatigue when so many other factors are changing. It can be noted however, that the increased RMS amplitude in the EMG occurred in spite of a reduction in muscle force of approximately 36 N (pre-fatigue: 3529.4 ± 483.9 N; post-fatigue 3493.5 ± 484.6 N). This fact, along with the agreement in the spectral composition and the RPE scores provide evidence that the subjects were fatigued.

The model used in this study produced results consistent with previous research. Lifting research often focuses on L5/S1 moments rather than L5/S1 compressive forces. Schipplein et al. (1990) showed peak moments at L5/S1 to be about 245 N•m while lifting a 10 kg load. de Looze et al. (1992) reported peak moment values at L5/S1 ranging from 174 to 257 N•m for a 18.8 kg load. In the present study, moment values at the time of the peak compressive forces at L5/S1 ranged from 162 to 267 N•m while lifting a10 kg load during the pre-fatigue condition. Chen (2000) focused on the peak compressive forces at L5/S1 and found that lifting a 10 kg load produced an average value of 3690 N for subjects with an average mass of 67 kg. The mean peak compressive force at L5/S1 for subjects in the current study (average mass: 77 kg) was 3820 N while lifting a 10 kg load during the pre-fatigue condition. Subjects did not show a statistically significant change in peak compressive force during fatigue but did show some changes in kinematics during the time of the peak compressive force. Accelerations of the load and the trunk increased with fatigue. Other studies have also shown increased accelerations during fatigue (Bonato, et al., 2002, Chen, 2000). We hypothesize that these increased accelerations are due to an attempt to compensate for decreased muscle force production by recruitment of higher threshold motor units. The motor unit pool during the pre-fatigue condition will be composed of relatively low threshold motor units when lifting a light load. As the force producing ability of these motor units decreases additional motor units must be recruited. The motor unit pool during fatigue will then be composed of motor units with a higher threshold: they will have a greater number of fibers per motor unit and thus less precision: and they will have a greater rate of tension development. It is therefore likely that the motor units in the latter stages of fatigue produce greater accelerations than those in pre-fatigued muscle. The load also showed greater displacement prior to reaching the peak compressive force during the fatigue. The greater peak accelerations were likely responsible for these results. The greater accelerations imply greater velocity changes, which imply greater position changes.

Increasing the load acceleration and the angular acceleration of the trunk should have increased the peak compressive force, however 8 of the 13 subjects actually decreased peak compressive forces during fatigue. There are several reasons why this occurred. First, the subjects that decreased peak compressive forces during fatigue showed lower magnitude increases in the load acceleration (0.4 m•s^{-2} vs 1.3 m•s^{-2}) and trunk acceleration (56.8 deg•s^{-2} vs 93.2 deg•s^{-2}). Second, these subjects also tended to increase the sacral angle during fatigue. This does not decrease spinal loading, but it does shift the loading from a compressive force to a shear force. Third, these subjects were better able to maintain posture during fatigue. The subjects that increased peak compressive forces began lifting in more of a stooped posture and had a greater shift toward even more of a stooped posture when they became fatigued. This shift toward a stooped posture is a common adaptation during fatigue and has been associated with increased compressive forces (Trafimow, et al., 1993, Chen, 2000).

Conclusions

Subjects that increased peak compressive forces when fatigued used more of a stooped posture. Weight training could be done to develop greater fatigue resistance in the muscles of the lower back to delay lifters from shifting to a more stooped posture. Subjects also increased the accelerations of the load and thus "jerked" the load off the ground during fatigue. Some of the subjects were able to decrease peak compressive forces while other subjects showed an increase in peak compressive forces. It was hypothesized that the increased accelerations were caused by a shift from the small fatigue resistant motor units used during the initiation of lifting to the large, lower-precision motor units used during fatigue. If this is the case, then appropriate training of the small motor units should delay the onset of fatigue and thus improve lifting performance.

Other methods that could help prevent the effects of repeated lifts are alternating the location of muscle stresses and distributing the muscle stresses over many joints. Fogleman and Smith (1995) and Resnick (1996) suggested variation in the strategies used by different individuals during lifting tasks. So, it is suggested that the strategy of distributing stresses among joints may also be a proper method of protecting the other joints from injury. Distribution of the muscle stresses over joints could be accomplished by using the squat technique, which may distribute the stresses over more joints than the stoop technique and thus reduce the peak stresses. In other words, individuals should lift with the entire body rather than only with specific joints.

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Table 3.1: Peak compressive forces at L5/S1 joint, kinematic and kinetic variables at the values of the peak compressive forces, from all subjects.

Variables	**Pre Fatigue**		**Post Fatigue**		**p**
	Mean	Stdev	Mean	Stdev	value
Peak Compressive Forces (N)	3821.9	± 484.9	3769.6	± 468.4	0.543
Vertical acceleration of the load ($m \cdot s^{-2}$)	2.5	± 2.2	3.2	± 2.9	0.004
Angular acceleration of the trunk ($deg \cdot s^{-2}$)	138.7	± 74.3	209.6	± 115.4	0.005
Horizontal distance between load and L5/S1 (m)	0.57	± 0.04	0.54	± 0.04	0.002
Height of the load (m)	0.35	± 0.04	0.37	± 0.07	0.037
Sacral angle (deg)	55.4	± 15.5	57.8	± 12.6	0.417
Postural Index	0.59	± 0.18	0.57	± 0.18	0.415

Table 3.2: Mean values from 10 pre fatigue and 10 post fatigue lifts of the 5 subjects that increased their peak compressive forces. Kinematic and kinetic variables were the values at the peak compressive forces.

Variables	**Pre Fatigue**		**Post Fatigue**	
	Mean	Stdev	Mean	Stdev
Peak Compressive Forces (N)	3664.3	± 195.3	3935.3	± 271.5
Vertical acceleration of the load ($m \cdot s^{-2}$)	2.4	± 0.6	3.7	± 0.8
Angular acceleration of the trunk ($deg \cdot s^{-2}$)	196.3	± 73.4	289.5	± 120.9
Horizontal distance between load and L5/S1 (m)	0.56	± 0.04	0.52	± 0.01
Height of the load (m)	0.36	± 0.02	0.41	± 0.07
Sacral angle (deg)	63.0	± 17.6	62.4	± 15.4
Postural Index	0.50	± 0.22	0.45	± 0.21

Table 3.3: Mean values from 10 pre fatigue and 10 post fatigue lifts of the 8 subjects that decreased their peak compressive forces. Kinematic and kinetic variables were the values at the peak compressive forces.

Variables	**Pre Fatigue**		**Post Fatigue**	
	Mean	Stdev	Mean	Stdev
Peak Compressive Forces (N)	3920.4	± 593.6	3666.0	± 549.5
Vertical acceleration of the load ($m•s^{-2}$)	2.5	± 0.6	2.9	± 0.7
Angular acceleration of the trunk ($deg•s^{-2}$)	102.8	± 50.8	159.6	± 83.7
Horizontal distance between load and L5/S1 (m)	0.57	± 0.04	0.55	± 0.04
Height of the load (m)	0.34	± 0.01	0.35	± 0.03
Sacral angle (deg)	50.6	± 12.7	55.0	± 10.4
Postural Index	0.65	± 0.14	0.64	± 0.14

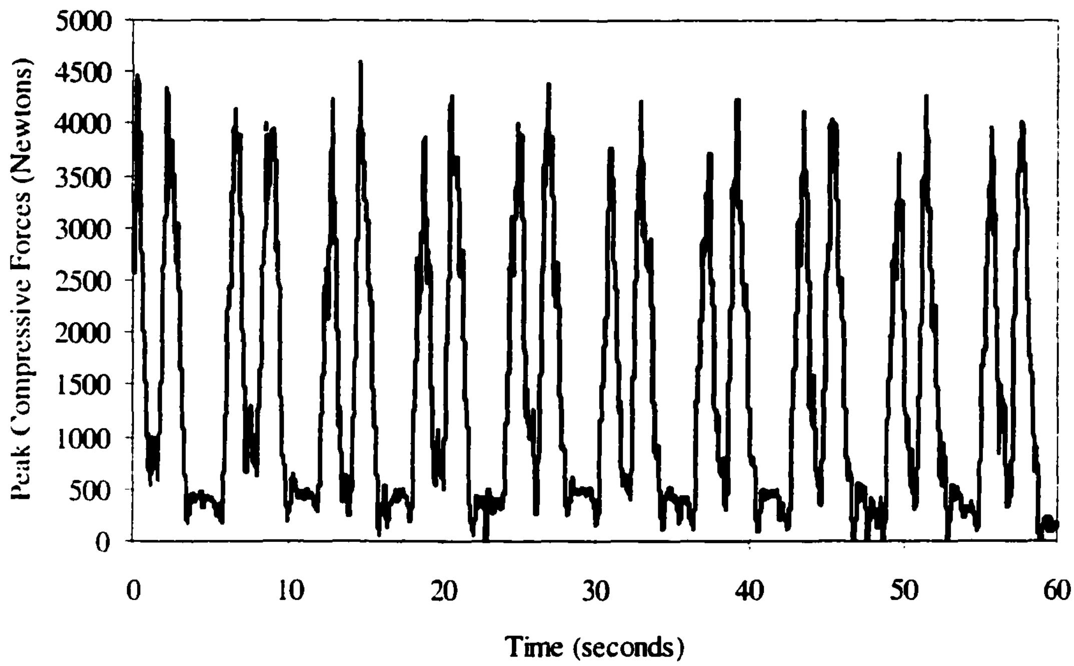

Figure 3.1: Example of peak compressive forces at L5/S1 joint during pre fatigue condition.

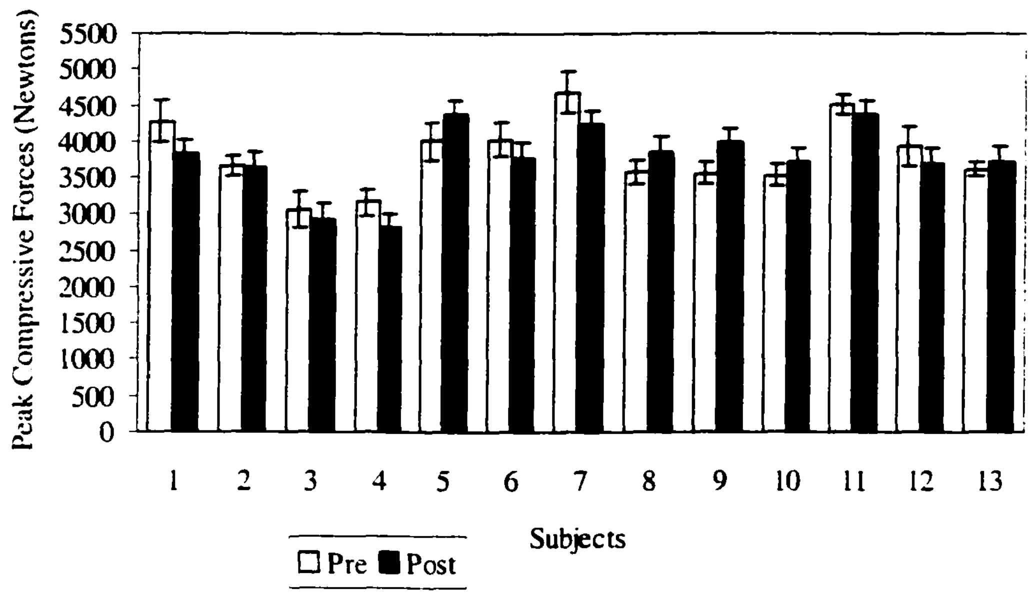

Figure 3.2: The peak compressive forces at L5/S1 joint for all subjects between the pre and post fatigue conditions.

CHAPTER 4: INFLUENCE OF PREVIOUS LIFTS AND KNOWLEDGE OF LOAD MAGNITUDE ON L5/S1 COMPRESSIVE FORCE DURING MANUAL LIFTING

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Abstract

This study investigated stresses on the lower back and adaptations of the lifts as effects of magnitude and knowledge of loads. Twenty male subjects performed four series of lifts on a container supporting either a light load (3 kg) or heavy load (17 kg). In two of the series the subjects knew the mass of the load and in another two they did not. When the light load was unknown, subjects overestimated the load, increasing the peak compressive forces at L5/S1 significantly. This increase was the result of an increased vertical acceleration of load and a farther displacement of load from L5/S1. Knowledge of the mass did not significantly alter the peak compressive forces while lifting a heavy load, but there was a reduction in vertical acceleration of load when load was unknown as subjects underestimated the load. Subjects were able to adapt when lifting an unknown load and the greatest adaptation occurred between the initial lift and subsequent lifts. Regardless of the knowledge of load, subjects adapted to the light load by reducing the stresses on the lower back. The subjects brought the load closer to lower back after the initial lift and overestimated the load during the unknown condition. When a heavy load was unknown, subjects adapted their lifts by increasing the stresses on lower back by a tendency to move the load farther from the lower back and increasing the vertical acceleration of load due to the underestimation of the load. When lifting a heavy load under both conditions, subjects de-emphasized the use of the back and increased the use of their legs.

Relevance to industry

The actual lifting activities involve manual handling of loads with unknown mass. This study investigated stresses on the lower back and adaptations of lift technique based on load magnitude and knowledge of the loads. The results provide meaningful information for lifting job design and the prevention of lower back injuries.

Keywords: lifting, compressive forces, load knowledge, adapt, subsequent lifts.

Introduction

The high occurrence of lower back injuries has developed into a serious health problem in industry. Manual material handling, especially lifting loads, is the most frequently reported cause of the lower back injuries (Chaffin and Park, 1973, Klein et al., 1984). This is because the act of lifting generates high mechanical stresses on musculoskeletal structures in the lower back. In some lifting jobs, such as refuse collecting and luggage dispatching, the workers often have no knowledge of the load magnitude prior to the actual lift. While starting a lift, workers can sometimes anticipate the necessary effort based on previous experience. However, this is not always possible because the weight of load is often unknown. When lifting an unexpectedly light load the initial effort may be too large. This results in a tendency to move the body and load upward rapidly in an uncontrolled manner (Butler et al., 1993, Patterson et al., 1987). Conversely, a heavy load can be underestimated, causing inappropriate body positioning resulted the development of unsafe forces in the body. Epidemiological studies have shown that workers exerting sudden unexpected maximal efforts are particularly vulnerable to low back disorders (Magora, 1973).

The evaluation of manual material handling tasks is often used to reduce the risk of low back injury at the workplace. The National Institute for Occupational Safety and Health (NIOSH, 1981) provides a method for computing a weight limit for manual lifting, in which several load-determining factors are involved. The factors are load location, the displacement of the load, the asymmetric angle of lifting, lifting frequency, and the coupling between the load and hands. However, load knowledge is a factor that is not accounted for in this method or in any other evaluation approach.

Previous research has shown that knowledge of load both known and unknown can influence lifting kinematics and the peak moment and forces at lower back during manual lifting tasks. Patterson et al. (1987) studied the effect of load knowledge on stresses at the L4/L5 joint in lifting of three different loads (6.8, 10.2, and 13.8 kg). They observed a general tendency towards higher peak moments in the condition when the actual mass of the load was unknown. This study also compared the effect of load knowledge to different types of lifters, experienced and novice lifters. The results showed that the experienced lifters had lower stress levels at L4/L5 than did the novice lifters due to the lifting technique.

Butler et al. (1993) investigated the peak moments at the L5/S1 joint as the function of two different lifting conditions: with and without knowledge of four different weights (no weight, 150, 250 and 300 N). The peak moment at L5/S1 joint for the 0-N (light load) condition was increased significantly in the without knowledge condition as compared to the with knowledge condition. At the other weights, no significant difference was found, except at 150 N where a significant increase in the speed of trunk extension under the unknown weight condition occurred. This study was different from Patterson et al. (1987) due to the magnitude of the loads used, comparing an extremely light (0 N) weight to heavy (300 N) weight.

de Looze et al. (2000) analyzed the low back moments in lifting as an effect of absence of load knowledge. Subjects in this study were only informed of the range of masses (6.5 – 16.5 kg) to be lifted for the unknown condition. The results found the peak L5/S1 moments in the unknown condition were significantly increased than in the known condition (actual mass to be lifted). The difference of this study from previous studies was the definition of unknown condition. Previous studies defined the unknown condition as a condition that subjects did not know the magnitude of the mass when lifting the load. In this study, the unknown condition was defined as condition in which subjects knew only the range of masses, not the actual mass.

These researches investigated the effect of change an unexpected load caused after performed a series of lifts with an expected load during lifting. Actually, the unexpected load condition was different from the unknown load condition. The unknown load condition during lifting was a condition in which individuals did not know the mass of load to lift. The unexpected load condition during lifting was the condition in which individuals did not know when the expected load being lifted would unexpectedly be changed. The unknown load condition seemed to create more realistic lifting tasks than did the unexpected load condition.

Commissaris and Toussaint (1997) studied lifting tasks having a sudden change of light mass (6 kg) after a long series of a heavy mass lifting (16 kg). The study found that the moment at L5/S1 joint was significantly different (a decrease) between lifting both masses until 150 ms after the box lift-off from the ground.

der Burg and Dieen (2001) studied the lifting tasks in which the load was unexpectedly changed. Subjects performed a series of lifting movements of light loads (either 1.6 or 6.6 kg), then suddenly a mass of 10 kg heavier (either 11.6 or 16.6 kg) was presented. The peak L5/S1 moments and peak compressive forces were not significantly different when the subjects were lifting either an expected (light load) or an unexpectedly heavier load. The difference of this study from the previous study is that the previous study inserted a lighter unexpected mass during a series of lifts of heavy mass, while this study inserted a heavier unexpected mass during a series of lifts of light mass.

In all of these studies, subjects performed a single lift for each condition, known and unknown load magnitude. The kinematics and kinetics differences between conditions were then analyzed. None of these studies the investigated the changes that would occur to subsequent lifts as an effect of load knowledge when lifting either a light or heavy load. Almost all of these studies found that lifting an object without knowing the magnitude of the load tended to result in increase lumbosacral stresses. These studies also showed the knowledge of the load could be used as a factor to change lifting mechanics. Very little information is currently available on how subjects adapt to those changes during subsequent lifts for both lifting a known and an unknown load. Based on these situations, the current study is developed to answer the following questions:

1. Are the differences in compressive forces at L5/S1 the same between lifting of light load (3 kg) and heavy load (17 kg) under known and unknown conditions?
2. How do subjects adapt during subsequent lifts when the magnitude of light load (3 kg) was known or unknown?
3. How do subjects adapt during the subsequent lifts when the magnitude of heavy load (17 kg) was known or unknown?

Methods

Subjects

Twenty healthy young male subjects with no history of back injuries participated in this study. The mean age was 21 ± 1.9 years; body mass was 84 ± 13.9 kg; and body height was 1.8 ± 0.08 m. The subjects signed informed consent to participate in this study in accordance with university policy. Prior to the start of the study, subjects were familiarized with the experiment protocol.

Protocol

Each subject was asked to lift a load under 4 different conditions. Each condition consisted of a series of 5 lifts in which a light (3 kg) or a heavy (17 kg) load was lifted. The mass of the load was either known or unknown during the first lift of each series. Each condition was performed two times, for a total of 40 lifts. Before performing the lifting experiment, subjects practiced lifting each of the masses. During each series a lift was performed every 6 seconds based on an audible tone. The subjects were allowed as much as rest as needed between conditions.

The container had handles located 12 cm off the ground (33 cm length, 19 cm width, 12 cm height) and supported the 3 kg or 17 kg masses. The center of the container was positioned approximately 26 cm horizontally from the subject's ankles.

Each subject wore shoes during the lifting experiment. The right foot of each subject was placed on the force platform during the lifting experiment. Subjects began each series of lifts from the standing position, then instructed to bend down, grab the container and returned to a standing position as they lifted the container. The container was then returned to the initial position. Subjects were encouraged to perform lifts at a 'normal' speed and use a natural lifting technique.

Model

Lifting motion was captured using four Peak Video and Analog Motion Measurement System (Peak Performance Technologies, Inc., Englewood, CO) video cameras. The video camera system measured the three-dimensional position of nine reflective markers during the motion. The position of each marker was digitized at 120 Hz and low-pass filtered using a fourth-order (zero lag) Butterwoth filter using a 2 Hz cut-off frequency. The markers were placed on the second toe, the posterior point of the heel, the lateral malleolus, the midpoint of the lateral joint line of the knee, the glenohumeral joint, the pelvis (3 markers), and a additional marker on the container. (See a detailed explanation of the placement markers in pelvic area in chapter 3).

A strain-gage force platform (Advanced Mechanical Technology, Inc., Newton, MA model OR 6-6 2000) was used to determine the location, direction and magnitude of external ground reaction forces. The force platform signals were sampled at 120 Hz and synchronized with the kinematic data using the event and video control unit (Peak Performance Technologies, Inc., Englewood, CO).

A five-segment rigid body model was used to estimate internal forces. The segments used were the foot, leg, thigh, pelvis and one segment above the L5/S1 joint (head/arms/trunk). It was assumed that these segments were connected by hinge joints, namely the ankle, knee, hip, and L5/S1. Each segment was assumed to have a fixed-point mass and constant moment of inertia (See a detailed explanation of segment's location of center mass, moment inertia and segment's mass in chapter 3).

The reaction forces and moments at each joint were calculated using inverse dynamics for the standard link segment model (Winter, 1990). Newtonian equations of motion were applied to each individual segment starting at the foot. Reaction forces and joint moments were then estimated for the proximal end of each segment and the end of process was then repeated for the next proximal segment in the model. Equations of motion and the anthropometric model were implemented using a custom analysis program. (See a detailed explanation of the compressive forces at L5/S1 in chapter 3)

Data analysis

The independent variables were two load magnitudes (3 kg-light and 17 kg-heavy) and two levels of load knowledge (known and unknown). The main dependent variable was the peak compressive forces at the L5/S1 joint. The timing of the peak compressive forces was observed to occur just after the load left the ground. To further analyze the causes of increased or decreased peak compressive forces additional variables were calculated at the time of the peak compressive force. These variables included the sacral angle, the vertical acceleration of the load, the horizontal distance between the center of load and L5/S1 joint, and the postural index. The postural index was defined as the ratio of the knee angle to the sum of the hip and L5/S1 angles.

Means and standard deviations were calculated for each of the five lifts conditions. The interactions between load magnitude (light and heavy) and knowledge of the load (known and unknown) were tested using repeated measures analysis of variance. Helmert contrasts were utilized to test how the subjects adapted lifting style when the light or heavy load was lifted in the known or unknown condition. These contrasts were used to compare each lift to the mean of subsequent lifts using an alpha level of 0.05 as the threshold for detecting statistical differences.

Results

The statistical interaction test between magnitude (light and heavy) and knowledge of the load (known and unknown) was performed to determine if knowledge of the load magnitude had the same effect on lifting a light weight as it did on lifting a heavy weight. These tests were performed only on the first lift of each series because after the first lift the load magnitude was not truly unknown. There was a significant interaction in peak compressive forces between the mass of the load and knowledge of the load magnitude ($p = 0.004$). While lifting a light load, the peak compressive forces at L5/S1 (figure 1) increased when the load was unknown (3624.3 ± 886.8 N vs 3964.2 ± 1104.6 N). While lifting a heavy load the peak compressive forces at L5/S1 (figure 1) decreased when the load was unknown (4931.2 ± 1213.3 N vs 4770.4 ± 1000.9 N).

There were significant kinematic changes that occurred when the load magnitude was unknown (table 1). The vertical acceleration of the load was increased by 17% (3.5 ± 0.9 $m \cdot s^{-2}$ vs 4.1 ± 1.3 $m \cdot s^{-2}$) when the light load was unknown, but it was decreased by 19% (2.5 ± 1.3 $m \cdot s^{-2}$ vs 2.1 ± 0.9 $m \cdot s^{-2}$) when the heavy load was unknown. The load was held further from L5/S1 in the horizontal direction (0.49 ± 0.04 m vs 0.51 ± 0.04 m) when the light load was unknown. However, knowledge of the load did not appear to influence this horizontal distance while lifting the heavy load (0.45 ± 0.09 m vs 0.43 ± 0.09 m). The postural index was also not influenced by knowledge of the light load (0.52 ± 0.13 vs 0.54 ± 0.18), but decreased by 0.04 when the heavy load was unknown (0.51 ± 0.18 vs 0.47 ± 0.18). The sacral angle decreased by 2.7 degrees (56.8 ± 13.9 deg vs 54.1 ± 15.2 deg) when the light load was unknown, but there was no effect of load knowledge while lifting the heavy loads (55.0 ± 13.4 deg vs 57.3 ± 14.3 deg).

To analyze how subjects adapted during each series of 5 lifts, each lift was compared to the mean of subsequent lifts. When the light load was known the first lift was different from the others in peak compressive force and horizontal distance between the load and L5/S1 (table 2). In addition, the fourth lift was different from the fifth lift in the vertical acceleration of the load (table 2). When the light load was lifted, the peak compressive forces at L5/S1 (figure 2) of the first lift was greater than that of subsequent lifts. There were also some statistically significant kinematic changes that occurred during the lifts. The horizontal displacement of the load relative to L5/S1 of the first lift was increased by compared to the subsequent lifts. The only statistically significant difference in the vertical acceleration of the known light load was a decrease of 0.3 m•s^{-2} between the 4th and the 5th lift. The sacral angle and the postural index did not significantly change during the lifts.

When the light load was unknown the first lift was different from the others in peak compressive force, horizontal distance and vertical acceleration of the load (table 2). In addition, the third lift was different from subsequent lifts in peak compressive force with the second lift was different from subsequent lifts in the horizontal distance (table 2). The peak compressive forces at L5/S1 (figure 2) of the first lift were greater than subsequent lifts. This drop in peak compressive force was accompanied by kinematic changes in the lifts. These included a decrease in the horizontal distance between the load and L5/S1 of 0.03 m and a decrease in the vertical accelerations of the load of 1 m•s^{-2} between the first and second lifts (table 2). The sacral angle and the postural index did not significantly change during the lifts.

When the heavy load was known the peak compressive forces showed no significant differences between the first lift and subsequent lifts (figure 3). There were a couple of kinematic changes that occurred between the first lift and subsequent lifts. There were some statistically significant kinematic changes that occurred during the lifts (table 3). The postural index of the first lift was statistically smaller than the subsequent lifts. The sacral angle of the first lift was statistically greater than the subsequent lifts. However, the vertical acceleration of the load and the horizontal displacement of load relative to L5/S1 did not significantly change during the lifts.

When the heavy load was unknown the peak compressive forces, horizontal distance between the load and L5/S1, the vertical acceleration of the load, the sacral angle and the postural index showed statistically significant changes between the first lift and subsequent lifts. The peak compressive forces at L5/S1 of the first lift were less than the subsequent lifts (figure 3). The vertical acceleration of the load and the distance between the load and L5/S1 both decreased after the first lift. The sacral angle of the first lift was greater than the subsequent lifts and the postural index of the first lift was smaller than the subsequent lifts

Discussion

This study applied a dynamic biomechanical model of lifting to estimate changes in kinematic and kinetic data that occurred with the effects of load knowledge and previous lifts. The purpose of this study was to examine the mechanical stresses on the lower back when subjects were lifting a load (light or heavy) with known or unknown magnitude and to analyze how they adapted during a series of 5 lifts.

Lifting studies frequently focus on L5/S1 moments rather than L5/S1 compressive forces. de Looze et al. (1992) reported the mean peak moment at L5/S1 ranged from 220.1 ± 25.2 N•m for an 18.8-kg load and an average subject's mass of 76 kg. The present study showed the moment values at the time of the peak compressive forces at L5/S1 ranged from 194.5 ± 11.4 N•m while lifting a 3-kg load and ranged from 269.4 ± 13.0 N•m while lifting a 17-kg load during the known conditions. Chen (2000) focused on the peak compressive forces at L5/S1 and showed that lifting a 5-kg load generated an average value of 3300 ± 370 N, average value of 4490 ± 520 N while lifting a 15-kg load, and average value of 5050 ± 500 N while lifting a 20-kg load for subjects with an average mass of 67 kg. During the known conditions, the mean peak compressive forces at L5/S1 in the current study (average subject mass: 84 kg) were 3535 ± 951 N while lifting a 3-kg load and 4896 ± 1110 N while lifting a 17-kg load.

In this study. while lifting light load (3 kg) and the mass was unknown subjects increased the peak compressive forces at L5/S1 while lifting light load (3 kg). This result supports previous studies which found that lifting a load under the unknown condition produces increased stresses on the lower back. especially for a light load (Patterson et al.. 1987. Butler et al.. 1993). However. the knowledge of the mass did not significantly change the peak compressive forces at L5/S1 while lifting a heavy load (17 kg). A previous study found that. for heavy loads. the lumbo-sacral moments were not influenced appreciably by whether the load magnitude was known or not (Butler et al.. 1993). Thus. there were differences in the peak compressive forces at L5/S1 while subjects were lifting a light load or a heavy load under the known or unknown condition. This was exemplified by the significant interaction between the mass and knowledge of the mass (figure 1).

While preparing to lift with the unknown load (light and heavy). subjects had to estimate the magnitude of the load. If a light load were estimated to be heavier load. subjects would overestimate the load. If it was a heavy load. but subjects estimated it as a light load. they underestimate the load. When subjects overestimate the load. the lifting technique generates a greater than normal acceleration. Likewise an underestimated load would show reduced accelerations. When the light load was unknown. the vertical acceleration of the load was increased during the first trial. This result agrees with previous study (Butler et al., 1993) that lifting light unknown weights resulted in an overestimation of the load. When the heavy load was unknown. the vertical acceleration of the load was reduced. suggesting an underestimation of the load. However. the previous study (Butler et al.. 1993) found lifting of heavy loads (15. 25 and 30 kg) did not show significant changes in the maximum velocity of the load.

When overestimating the load. subjects generated a rapid lifting movement. The rapid movement produced an uncontrolled movement ("jerk"). a sudden pull on the load. with rapid acceleration of the load and the occurrence of the lumbar spine extension (Butler et al., 1993). The extension of the pelvis segment will reduce the sacral angle. In this study, while lifting the unknown light load the sacral angle was significantly reduced, agreeing with Butler et al. (1993). When the light load was unknown, the load was located farther from L5/S1. The greater accelerations were likely responsible for this result, as greater accelerations imply greater velocity changes, which imply greater position changes.

Previous studies have not considered how individuals adapt during subsequent lifts when lifting light or heavy loads for situations in which the mass is known or unknown. Typically only a single lift is investigated rather than a series of lifts. In the current study, subsequent lifts were analyzed to see how subjects adapt to lifting a light or a heavy load under the known and unknown conditions. Subjects adapted more when lifting an unknown load as compared to a known load. This is because subjects had no knowledge of the mass during the initial lift in the unknown condition. After that, subjects made kinematic and kinetic adaptations during subsequent lifts based on the experience of the first lift. In this current study, the greatest adaptations occurred between the initial lift and the subsequent lifts.

Regardless of the knowledge of the mass, subjects adapted to the light load by reducing the stresses on the lower back (table 2). This was because subjects tended to move the load closer to L5/S1 during the subsequent lifts regardless of knowing the mass of load. When the light load was unknown, greater reduction in the lower back stresses occurred. Part of this reduction was due to overestimating the load during the first lift of the unknown condition. Evidence for this overestimation was found in the increased vertical acceleration of the load during the first lift. Subjects did not adapt the lifting technique (postural index) during the light lifts. This was probably because the light mass did not require any postural adjustment during this short period of lifting. While lifting the light load there were few adaptations that occurred after the second lift (table 2).

When a heavy load was unknown, subjects adapted during subsequent lifts by increasing the stresses on the lower back. This was because subjects tended to move the load farther to L5/S1. Part of this increase was due to underestimating the load during the first lift of the unknown condition. The evidence of this underestimation could be found in the decreased vertical acceleration of the load during the first lift. While lifting a heavy load under known condition, subjects did not change the stresses on the lower back during the lifts. This was because although subjects tended to reduce the sacral angle during the lifts (table 3), they also shifted toward a squat technique. The squat (knee) technique has been shown to generate less biomechanical stresses on the lower back than the stoop (back) technique (Andersson and Chaffin, 1986). However, the reduction of sacral angle would increase the compressive forces on the lower back (Chaffin, et al., 1999). Regardless of the knowledge of the mass, subjects de-emphasized use of the back and increased use of the legs. This can be seen in the increase of the postural index (table 3).

Conclusion

The present study showed individuals lift differently when they knew and did not know the magnitude of the load being lifted. Knowing the magnitude of the load influenced both lifting a light and heavy load. While lifting the light load without knowing the magnitude, individuals tended to overestimate the load. The overestimation generated a rapid vertical acceleration of the load that increased the stresses on lower back. On other hand, while lifting the heavy load without knowing the magnitude, individuals tended to underestimate the load. The underestimation reduced the vertical acceleration of the load, decreasing the stresses on the lower back.

The current study also showed individuals made changes (adaptations) during the subsequent lifts. Individuals adapted the lifts more when lifting an unknown load. The greatest adaptations occurred between the initial and the subsequent lifts. Individuals adapted the lifts differently depending on the magnitude of the load. While lifting a light load, regardless of the knowledge of the mass, individuals adapted the subsequent lifts by reducing the stresses on the lower back. Regardless of the knowledge of the mass when lifting a heavy load, individuals adapted the subsequent lifts by increasing the use of the legs.

To prevent the risks of lower back injury due to the misinterpretations of load being lifted under the unknown condition (overestimate or underestimate), the information of the mass should be given as clear as possible. Providing a label/mark on the container that displays the mass (weight) is a good solution to inform people the magnitude of the load. If an exact amount of mass is not available, the approximately masses (range of the masses) of the loads should still be provided, so that people can estimate the magnitude of the loads before lifting.

People making changes (adaptations) that reduce the stresses on the lower back during the lifts, especially when lifting a heavy mass.

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Table 4.1: The first lifts (means ± standard deviations) of light (3 kg) and heavy (17 kg) load when the mass was known and unknown.

	Light known	Light unknown	Heavy known	Heavy unknown
Peak compressive forces (N)	3624.3 ± 886.8	3964.2 ± 1104.6	4931.2 ± 1213.3	4770.4 ± 1000.9
Horizontal distance between the load and L5/S1 (m)	0.49 ± 0.04	0.51 ± 0.04	0.45 ± 0.09	0.43 ± 0.09
Vertical acceleration of the load ($m \cdot s^{-2}$)	3.5 ± 0.9	4.1 ± 1.3	2.5 ± 1.3	2.1 ± 0.9
Sacral angle (deg)	56.8 ± 13.9	54.1 ± 15.2	55.0 ± 13.4	57.3 ± 14.3
Postural Index	0.52 ± 0.13	0.54 ± 0.18	0.51 ± 0.18	0.47 ± 0.18

Table 4.2: The five lifts (means ± standard deviations) of light (3 kg) load when the mass was known and unknown.

Light known	**1**	**2**	**3**	**4**	**5**	**Sig.***
Peak compressive forces (N)	3624.3 ± 886.8	3509.4 ± 944.5	3483.3 ± 946.3	3555.7 ± 953.9	3499.4 ± 1021.9	1
Horizontal distance between the load and L5/S1 (m)	0.49 ± 0.04	0.48 ± 0.04	0.47 ± 0.04	0.47 ± 0.04	0.48 ± 0.04	1
Vertical acceleration of the load (m•s^{-2})	3.5 ± 0.9	3.4 ± 0.9	3.3 ± 0.9	3.6 ± 0.9	3.3 ± 0.9	4
Sacral angle (deg)	56.8 ± 13.9	56.9 ± 15.2	57.4 ± 14.8	56.3 ± 15.2	55.6 ± 18.8	
Postural Index	0.52 ± 0.13	0.52 ± 0.18	0.51 ± 0.18	0.52 ± 0.18	0.52 ± 0.18	
Light unknown	**1**	**2**	**3**	**4**	**5**	**Sig.***
Peak compressive forces (N)	3964.2 ± 1104.6	3557.9 ± 1037.1	3606.9 ± 1058.6	3521.7 ± 1029.9	3491.1 ± 986.6	1, 3
Horizontal distance between the load and L5/S1 (m)	0.51 ± 0.04	0.48 ± 0.04	0.47 ± 0.04	0.46 ± 0.04	0.47 ± 0.04	1, 2
Vertical acceleration of the load (m•s^{-2})	4.1 ± 1.3	3.1 ± 0.9	3.4 ± 0.4	3.1 ± 0.9	3.2 ± 0.9	1
Sacral angle (deg)	54.1 ± 15.2	55.7 ± 15.7	56.0 ± 16.5	54.7 ± 15.7	56.2 ± 16.5	
Postural Index	0.54 ± 0.18	0.52 ± 0.18	0.53 ± 0.18	0.53 ± 0.18	0.53 ± 0.18	

* indicates that the identified lifts are significantly different from the subsequent lifts.

Table 4.3: The five lifts (means ± standard deviations) of heavy (17 kg) load when the mass was known and unknown.

Heavy known	**1**	**2**	**3**	**4**	**5**	**Sig.***
Peak compressive forces (N)	4931.2	4937.2	4870.5	4862.4	4876.7	
	± 1213.3	± 1088.5	± 1030.8	± 1135.5	± 1082.7	
Horizontal distance between	0.45	0.47	0.46	0.46	0.46	
the load and L5/S1 (m)	± 0.09	± 0.09	± 0.09	± 0.09	± 0.09	
Vertical acceleration of the load ($m \cdot s^{-2}$)	2.5	2.9	2.9	2.9	2.8	
	± 1.3	± 1.3	± 0.9	± 0.9	± 0.9	
Sacral angle (deg)	55.0	52.2	52.7	51.4	52.9	1
	± 13.4	± 13.9	± 13.9	± 14.3	± 15.7	
Postural Index	0.51	0.54	0.54	0.57	0.55	1
	± 0.18	± 0.18	± 0.18	± 0.18	± 0.18	
Heavy unknown	**1**	**2**	**3**	**4**	**5**	**Sig.***
Peak compressive forces (N)	4770.4	5025.9	4953.0	4887.9	4958.9	1
	± 1000.9	± 1102.8	± 1186.5	± 1139.9	± 1298.7	
Horizontal distance between	0.43	0.48	0.47	0.46	0.47	1
the load and L5/S1 (m)	± 0.09	± 0.09	± 0.09	± 0.09	± 0.09	
Vertical acceleration of the load ($m \cdot s^{-2}$)	2.1	2.9	2.8	2.8	2.8	1
	± 0.9	± 0.9	± 0.9	± 0.9	± 0.9	
Sacral angle (deg)	57.3	51.5	51.8	52.5	52.5	1
	± 14.3	± 15.7	± 14.8	± 13.9	± 16.1	
Postural Index	0.47	0.54	0.55	0.55	0.54	1
	± 0.18	± 0.18	± 0.18	± 0.18	± 0.18	

* indicates that the identified lifts are significantly different from the subsequent lifts.

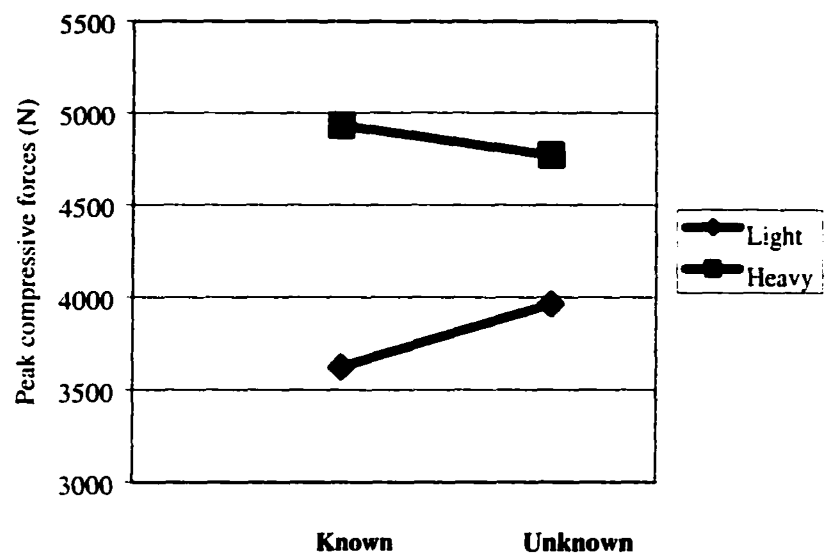

Figure 4.1: Peak compressive forces during the first lifts of light load (3 kg) and heavy load (17 kg) when the mass was known and unknown.

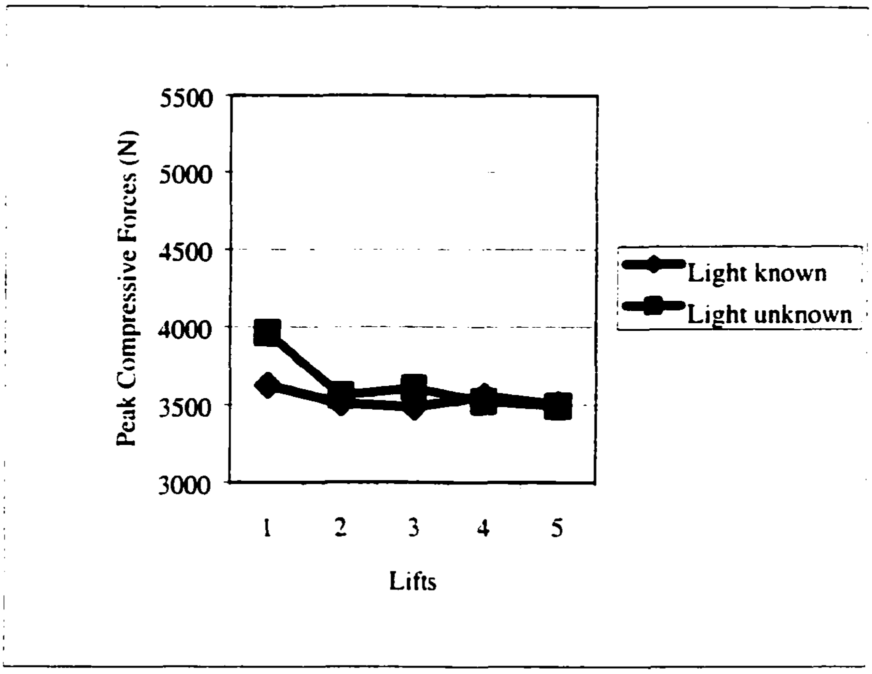

Figure 4.2: Peak compressive forces of the five lifts of light load (3 kg) known and unknown conditions.

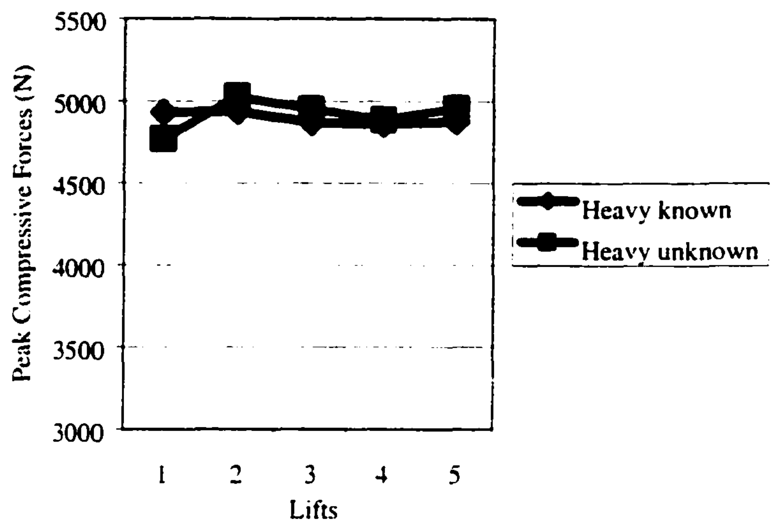

Figure 4.3: Peak compressive forces of the five lifts of the heavy load (17 kg) known and unknown conditions.

CHAPTER 5: INFLUENCE OF CONSTRAINING BARRIER ON L5/S1 COMPRESSIVE FORCE DURING MANUAL LIFTING

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Abstract

The purpose of this study was to examine the mechanical stresses on the lower back as the response of different heights of constraining barrier. Ten male subjects lifted a load from the floor to the knuckle height under the non-constrained and the constrained conditions with 4 different heights of constraining barrier (0%, 80%, 100%, 120% and 140% of knee height). The constrained condition was defined as the condition where a load was placed on the floor behind a certain level of bar. When lifting of the constrained conditions, subjects significantly increased the peak compressive forces at L5/S1 compared to the non-constrained (3868.8 ± 527.5 N, 4175.0 ± 486.0 N, 4162.4 ± 462.3 N, 4136.0 ± 553.1 N, 4079.4 ± 468.9 N for 0%, 80%, 100%, 120% and 140% barrier height conditions respectively). The subjects moved the load further from L5/S1 in the horizontal direction when lifting during the constrained conditions. While lifting during the constrained conditions subjects generated an increase in the sacral angle and a decrease of the knee flexion. In this study, the peak compressive forces at L5/S1 showed a statistically significant quadratic trend. However, the magnitude of the difference of peak compressive forces during the constrained conditions was small. Future research and more data collection of lifting tasks for different heights of barrier are required to emphasize the meaning of the quadratic trend of compressive forces.

Relevance to industry

Lifting load over a constraint barrier has implications for many manual material handling tasks in industry. The constrained condition increased the peak compressive forces in the lower back during lifting due to an increased load displacement and a decreased of the knee flexion.

Keywords: lifting, lower back injury, compressive forces, constraint barrier.

Introduction

Epidemiologic research indicates that lower back pain is a major problem in terms of both humans suffering and cost for workers. Manual material handling tasks, especially repetitive lifting, is most commonly reported as the cause of back injuries (Chaffin and Park, 1973, Frymoyer et al., 1983, Klein et al., 1984). This is because the repetitive lifting produces high compressive stresses on the back, especially the lower back (Dolan et al., 1994), which can cause the degeneration on the annulus fibrosus of the intervertebral discs and then cause the lumbar disc to prolapse posteriorly (Adams and Hutton, 1983, 1985).

Researchers have developed a considerable interest in how people apply the motion and strategy of lifting and how people control the effect of lifting on the human body (NIOSH, 1981). These studies that have focused on the stress occurring in the lower back have utilized a biomechanical model (de Looze, et al., 1992, Schipplein et al., 1990, Tsuang et al., 1992). In this approach, the compressive forces acting on the lower back are estimated based on the reactive forces at the L5/S1 intervertebral disc center, knowledge of musculature and the sacral orientation. The models assume rigid body segments connected by hinge joints. The magnitude of loads being lifted and anthropometric measurements of body segments are required along with knowledge of external forces acting on the body. Applying conventional Newtonian equations of motion, the joint reactive forces and moments are predicted and L5/S1 compressive forces are calculated.

Individuals sometimes require lifting from, and lowering into, industrial bins. These bins form a constraint barrier that can serve to restrict the preferred motion of the body and result in altered stresses. Lifting tasks with different barrier heights can influence lifting kinematics, moments, and compressive forces at L5/S1 (the joint between the fifth lumbar and the first sacral vertebrae) during manual lifting. McKean and Potvin (2001) studied the effects of a simulated industrial bin on lifting and lowering posture and trunk muscle activity (EMG). The wall height of the simulated industrial bin was constructed at 120% of the average male/female knee height. Male subjects lifted a 15 kg load and female subjects lifted a 8.5 kg load. Each subject performed 10 lifts and 10 lowers under both "freestyle" and "constrained" conditions. The load was generally lifted and lowered at a greater peak horizontal displacement for the constrained condition than the freestyle condition. Subjects tended to have larger peak pelvis and trunk flexion angles, but less peak knee angle for the constrained condition than for the freestyle condition. The peak EMG magnitudes for both thoracic and lumbar muscle groups also were higher for the constrained than for the freestyle condition.

The previous study was one of the only studies to look at the effects of a constraining barrier on lifting and lowering, even though it has implications to many material handling tasks applied in industry. This study did not directly calculate the mechanical stresses (i.e., moments, compressive forces) that occurred on the lower back in response to a constraining barrier. Changes in the magnitude of back muscle activity (EMG) were used to represent the loading of the back. In addition this study compared a single constrained condition (120% of knee height) to the non-constrained condition.

The purpose of this study was to examine the mechanical stresses on the lower back in response to the different levels of constraining barrier. Lifting tasks over a non-constraining barrier and four different heights of constraining barrier were performed. The heights of the constraining barrier were constructed to be 0%, 80%, 100%, 120% and 140% of knee height.

Methods

Subjects

Ten healthy young male subjects with no history of back injuries participated in this study. The mean age was 25 ± 4.9 years; body mass was 75 ± 8.8 kg; and body height was 1.75 ± 0.06 m. The subjects signed informed consent to participate in this study in accordance with university policy. Prior to the start of the study, subjects were familiarized with the experiment protocol.

Protocol

Each subject was asked to lift a load under 5 different conditions. Each condition consisted of a series of 10 lifts. The conditions were the non-constrained (as a 'free-normal' lifting) condition and the constrained conditions, with 4 different heights of constraining barrier. The constrained condition was defined as the condition where the load was placed on the floor behind a certain height of constraining barrier. The constraining barrier was a bar that placed between the subject and the load. The heights of the constraining barrier were constructed to be 0%, 80%, 100%, 120% and 140% of the average male knee height. The knee height was calculated as 28.5% of total body height (Chaffin and Andersson, 1999). The order of presentation of these conditions was balanced for each subject. During each condition a lift was performed every 6 seconds based on an audible tone. The subjects were permitted as much as rest as needed between conditions.

The load was a crate (42 cm length, 34.5 cm width, 27 cm height) that contained a mass of 10.3 kg placed on the floor in front of the subject. The crate had two fixed handles placed symmetrically 27 cm above the bottom. The handles and mass center of the crate were positioned approximately 27 cm horizontally from the subject's ankles.

Each subject wore shoes during the experiment. The right foot of each subject was placed on the force platform during the experiment. Subjects began the lifting experiment from the standing position, then were asked to bend down, grab the crate and returned to a standing position as they lifted the crate. The crate was then returned to the original position. Subjects were encouraged to perform lifts at a 'normal' speed and use a natural lifting technique.

Model

The motions in this study were recorded using four 120 Hz video cameras (Peak Performance Technologies, Inc., Englewood, CO). The three-dimensional positions of nine reflective markers were recorded. Reflective markers were placed on the second toe, the posterior point of the heel, the lateral malleolus, the midpoint of the lateral joint line of the knee, and the glenohumeral joint. Three additional markers were then placed on the pelvis region to help defining the joint between the fifth lumbar and first sacral vertebrae (L5/S1). (See a detailed explanation of L5/S1 joint definition in chapter 3). A final marker was placed on the crate. The marker coordinates were low-pass filtered with a fourth-order (zero lag) Butterwoth filter using a 2 Hz cut-off frequency.

A strain-gage force platform (Advanced Mechanical Technology, Inc., Newton, MA model OR 6-6 2000) was used to determine the three components of ground reaction forces, the location, direction and magnitude of external forces. The force platform signals were sampled at 120 Hz and synchronized with the kinematic data using the event and video control unit (Peak Performance Technologies, Inc., Englewood, CO).

The body was modeled with five segment (foot, leg, thigh, pelvis, and a segment of head/arms/trunk) with intersegmental joints at the ankle, knee, hip, and L5/S1. Each segment was assumed to have a fixed point mass and constant moment of inertia (See a detailed explanation of segment's location of center mass, moment inertia, and segment's mass in chapter 3).

The inverse dynamics of the standard link segment model (Winter, 1990) and Newtonian equations of motion starting at the foot were applied to determine the reaction forces and moments at the proximal end of each segment. (See a detailed explanation of the compressive forces at L5/S1 in chapter 3).

Data analysis

The independent variable was the barrier height with 5 levels. The main dependent variable was the peak compressive forces at the L5/S1 joint. The time of the peak compressive forces was observed slightly after the time that subjects lift-off the load. In order to determine the cause of differences in the peak compressive forces, several variables were examined at the time of peak compressive forces. These variables included the horizontal distance between the center of load and L5/S1 joint, the vertical acceleration of the load, the angular acceleration of the trunk, the sacral and trunk angles and the postural index. The sacral angle was defined as the angle formed by the base of sacrum (Chaffin, et al., 1999). It was assumed to be 45° during static trial (Thieme, 1950). The postural index was defined as the ratio of the knee angle to the sum of hip and L5/S1 angles.

Means and standard deviations were calculated for each condition. Repeated measures analysis of variance was performed to detect statistical differences between the conditions. In addition, a polynomial regression model of second order was used to analyze the trend in the peak compressive forces after adjusting for the unequal intervals between conditions. An alpha level of 0.05 was selected as the level of statistical significance.

Results

There was a statistically significant increased in peak compressive forces at L5/S1 (figure 1) between the non-constrained and the constrained conditions ($p < 0.05$). Compared to the 0% barrier height the peak compressive forces at L5/S1 were increased by 306.2 N (7.9%), 293.6 N (7.6%), 267.2 N (6.9%) and 210.6 N (5.4%) for the 80%, 100%, 120% and 140% barrier heights respectively (table 1). The analysis of the trend of the peak compressive forces at L5/S1 showed statistically significant quadratic trend (p = 0.0006). However, the magnitude of the difference of the peak compressive forces during the constrained conditions was small. The peak compressive forces difference between the lowest (80%) and the highest (140%) barrier heights was less than 100 N.

There were significant kinematic changes that occurred, mostly between the 0% barrier height and the other barrier heights (table 1). The load was moved further from L5/S1 when the barrier was introduced. Compared to the 0% barrier height the load was moved further from L5/S1 in the horizontal direction by 3 cm, 4 cm, 3 cm and 2 cm for the 80%, 100%, 120% and 140% barrier heights respectively.

The forward tilt of the pelvis increased as the barrier height increased. This caused an increase in the sacral angle of 3.7 degrees, 5.2 degrees, 5.9 degrees and 8.2 degrees for the 80%, 100%, 120% and 140% barrier heights respectively (table 1).

There was a trend to decrease the postural index as the barrier height increased. Compared to the 0% barrier height the postural index decreased by 0.07, 0.09, 0.08 and 0.11 for the 80%, 100%, 120% and 140% barrier heights respectively (table 1). The decrease in the postural index was mainly due to the reduction of the knee angle.

The trunk angle, the angular acceleration of the trunk and the vertical acceleration of the load showed no statistically significant changes.

Discussion

The current study applied a dynamic biomechanical model to estimate the mechanical stresses and kinematic changes that occurred as a result of lifting a load over a constraining barrier. Peak compressive forces and moments were consistent with previous studies. de Looze et al. (1992) reported peak moment values at L5/S1 ranged from 220.1 ± 25.2 N•m for a 18.8 kg load. In the present study, moment values at L5/S1 ranged from 215.9 ± 9.4 N•m while lifting a 10.3 kg load during the non-constrained condition. Chen (2000) focused on the peak compressive forces at L5/S1 and found that lifting a 10 kg load produced an average value of 3690 ± 410 N for subjects with an average mass of 67 kg. The mean peak compressive forces at L5/S1 for subjects in the current study (average mass: 75 kg) was 3869 ± 528 N while lifting a 10.3 kg load during the non-constrained condition.

Subjects significantly increased peak compressive forces at L5/S1 when they had to lift the load over a barrier. This was because subjects moved the load further from L5/S1 in the horizontal direction when lifting during the constrained conditions. The greater displacement of the load relative to lower back generates the larger stresses at the lower back. This result supports the previous study (McKean and Potvin, 2001), which showed that lifting over the constraining barrier produced larger horizontal distances from hands to ankles than lifting under the freestyle (no barrier) condition. The larger distance contributes to increased the extensor moment demands and this was confirmed by the increased erector spinae muscle activity.

Increasing barrier height resulted in an increase of the sacral angle and decrease of the postural index. The increase in the sacral angle was due to an increase in forward pelvic tilt. The decrease in the postural index was mainly due to the reduction in the knee angle. These results agree with McKean and Potvin (2001) who found lifting over the constrained condition significantly reduced the amount of flexion at the knees and increased the peak trunk flexion compared to the freestyle condition. Potvin et al. (1991) and McKean and Potvin (2001) discovered the main contributor to increases in the peak trunk flexion would come from the pelvic flexion.

There was a significant quadratic trend in the peak compressive forces. Subjects tended to produce greater peak compressive forces at L5/S1 when lifting over the constraining barriers, where the height was close to knee height (80% - 100%) compared to the constraining barriers, where the height was higher than knee height (120% - 140%) or lower than knee height (0%). This situation prevented subjects from bringing the load closer to the body and thus greater peak compressive forces were generated in the lower back. In addition, the trajectory of the load needs to be more vertical when the load must be lifted over the barrier. When lifting over the higher barrier than knee height, subjects increased the forward pelvic tilt and might move the load closer to the body. An increase of the forward pelvic tilt means a greater sacral angle. The greater sacral angle does not decrease the spinal loading, but it does a shift the loading from compressive to shear.

Conclusions

In the current study, when lifting over a constraint barrier, individuals significantly increased the peak compressive forces in the lower back. This was because the constrained condition produced greater horizontal displacement of the load to the lower back than did the non-constrained. Lifting load over a constraint barrier also generated an increase of sacral angle and a decrease of the knee flexion.

The peak compressive forces in the lower back in this study showed a quadratic trend. Therefore, individuals tended to produce greater stresses in the lower back while lifting a load over constraining barrier where the height was close to knee height compared to a constraining barrier where the height was higher or lower than knee height. However, the magnitude of the difference of the peak compressive forces was very small. Future research should investigate barrier height between 0% and 80% and above 140% of knee height.

To prevent the risks of the lower back injury, people should avoid lifting load over a barrier if possible. This is because the constrained condition significantly generates a larger horizontal displacement between the load and the lower back, which will produce greater stresses on the lower back. So, when lifting people should keep the load close to the body. Lifting load under the constrained condition requires that people use more back than knees. This situation also develops more stresses on the lower back. If a barrier is necessary, it my be marginally beneficial to avoid barrier heights that restrict knee flexion.

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Table 5.1: Means and standard deviations of the peak compressive forces at the L5/S1 and other variables from all conditions. All variables were measured at the time of the peak compressive forces.

Variables	0% Cond 1	80% Cond 2	100% Cond 3	120% Cond 4	140% Cond 5	Sig.
Peak Compressive Forces (N)	**3868.8** ± 527.5	4175.0 ± 486.0	4162.4 ± 462.3	4136.0 ± 553.1	4079.4 ± 468.9	1 vs 2, 3, 4, 5
Horizontal distance between load and L5/S1 (m)	0.75 ± 0.38	0.78 ± 0.38	0.79 ± 0.44	0.78 ± 0.38	0.77 ± 0.32	1 vs 2, 3, 4, 5
Load vertical acceleration ($m \cdot s^{-2}$)	3.4 ± 0.9	3.6 ± 0.7	3.5 ± 0.5	3.7 ± 0.9	3.4 ± 0.6	none
Angular acceleration of the trunk ($deg \cdot s^{-2}$)	245.7 ± 111.6	275.7 ± 77.2	274.20 ± 80.9	260.9 ± 97.4	252.3 ± 83.8	none
Trunk angle (deg)	3.1 ± 12.3	0.2 ± 8.5	-0.2 ± 6.0	1.1 ± 7.6	0.5 ± 5.7	none
Sacral angle (deg)	62.4 ± 13.3	66.1 ± 10.4	67.6 ± 9.8	68.3 ± 10.4	70.6 ± 9.5	1 vs 2, 3, 5 5 vs 2, 3
Knee angle (deg)	54.3 ± 8.4	50.9 ± 6.2	48.0 ± 5.1	48.6 ± 5.9	43.9 ± 4.8	1 vs 2, 3, 4, 5 2 vs 5
Postural Index	0.47 ± 0.19	0.40 ± 0.16	0.38 ± 0.13	0.39 ± 0.13	0.36 ± 0.13	1 vs 2, 3, 4, 5 2 vs 5

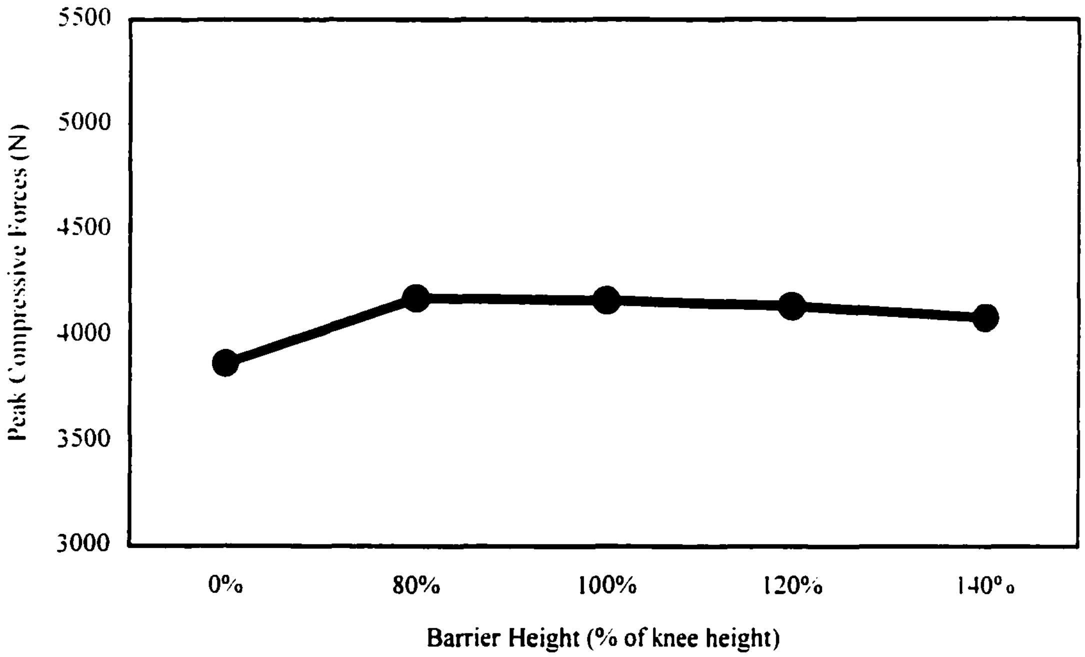

Figure 5.1: Peak compressive forces at the L5/S1 for all conditions.

CHAPTER 6: CONCLUSIONS

Lower back injury is a problem in the human society and manual material handling (MMH), especially the manual lifting task, has been thought to be the major cause of this type of injury.

This research was undertaken to investigate these types of motions. The specific purpose of this research was to examine the mechanical stresses at the lower back (the L5/S1 joint) and to analyze how individuals adapted to the stresses during dynamic manual lifting. A biomechanical model of lifting was applied to measure the stresses at the lower back and monitor kinematic changes. This dissertation contained three studies of lifting with three different aspects that caused the mechanical stresses in the lower back. The first study investigated the effects that fatigue had on the stresses at the lower back. In this study, subjects performed a continuously lifting task until they were fatigued. The magnitude of muscular fatigue was documented by the changes in the median frequency and RMS of the electromyography. The rating of perceived exertion (RPE) was also used to monitor the overall discomfort during the lifts. The second study explored the effects of knowledge of load magnitude on the stresses at the lower back and also analyzed how subjects adapted to the stresses during subsequent lifts. In this study, subjects performed lifting tasks of two load magnitudes (light and heavy) and two levels of load knowledge (known and unknown). Each lifting task consisted of a series of five lifts. The third study examined the effects of a constraining barrier on the stresses at the lower back. In this study, subjects performed lifting tasks over five different heights of constraining barrier. The constraining barrier was a bar that was placed between the subject and the load. The height of constraining barrier was constructed as a percentage of knee height.

Research contributions and suggestions

This research has been undertaken because of the prevalence of lower back injury and the relation to lifting activities as the major cause. Three different aspects that influenced the mechanical stresses at the lower back during lifting were analyzed separately in three different studies. A biomechanical approach was used to explain the results of these studies.

Fatigue has been discovered to influence the stresses at the lower back during repetitive lifting task. Sparto, et al. (1997) and Adam and Dolan (1998) found that fatigue decreased the stresses at the lower back. Chen (2000) found that fatigue increased the compressive forces at the lower back. However, the result of this research showed that the peak compressive forces at L5/S1 were insignificantly different when fatigue was reached. The contribution of this research is to understand why the stresses at the lower back were different when fatigue was reached. This was because each individual subject applied different lifting strategies when they were fatigued. Some subjects increased and others decreased the peak compressive forces when fatigued. When fatigued, the accelerations of load and trunk significantly increased. It is hypothesized that the increased accelerations were caused by a shift from the small fatigue resistant motor units used during the initiation of lifting to the large, lower precision motor units during fatigue. Increasing the vertical acceleration of the load and angular acceleration of the trunk should have increased the peak compressive forces at the lower back, however some subjects decreased the peak compressive forces during fatigue. This was because subjects that decreased the peak compressive forces showed smaller magnitude increases in the load acceleration and trunk acceleration. They also tended to increase the sacral angle and maintain posture when fatigued. On the other hand, subjects that increased the peak compressive forces shifted toward to a stooped (back) posture when they were fatigue. In order to prevent an increase of stresses at lower back as the effects of repeated lifts, it is suggested that people should lift with the entire body rather than only specific joints. In addition, people also should perform appropriate training exercise of the back to delay the onset of fatigue of the lower back.

The knowledge of the load magnitude has shown to influence the stresses at the lower back during lifting. Individuals lift differently when they know and do not know the magnitude of the load. Previous studies (Patterson et al., 1987, Butler et al., 1993, Commissaris and Toussaint, 1997, de

Looze et al., 2000) have found that the stresses at the lower back significantly increased when individuals lift a load under the unknown condition. The result of this research also showed that when subjects lifting a light load under the unknown condition, the peak compressive forces at L5/S1 significantly increased. This was because subjects overestimated the load this produced a rapid acceleration, which increased the stresses at the lower back. The contribution of this research is the previous studies have not considered how individuals adapt during subsequent lifts when lifting light or heavy loads when the mass is known or unknown. Typically only a single lift is investigated rather than a series of lifts. In the current research, subsequent lifts were analyzed to see how subjects adapt to lifting a light or a heavy load under the known and unknown conditions. The results showed that subjects adapted during subsequent lifts differently depend on the knowledge and magnitude of the load. When lifting an unknown load, there were greater adaptations as compared to a known load. This was because subjects had no knowledge of the mass during the initial lift of the unknown condition. After that, subjects made kinematic and kinetic adaptations during subsequent lifts based on the experience of the first lift. In this research, the greatest adaptations occurred between the initial lift and the subsequent lifts. When lifting a light load, regardless of the knowledge of the load, subjects adapted during the subsequent lifts by reducing the peak compressive forces at the lower back. However, when lifting a heavy load regardless of the knowledge of the load, subjects did not reduce the stresses at the lower back. They adapted during the subsequent lifts by increasing the use of the legs. In order to reduce the risks of lower back injury due to the misinterpretations of load under the unknown condition, it is suggested that subjects be given knowledge of the mass of the load being lifted as clearly as possible. Providing a label/mark on the container that displays the mass (weight) is a good solution to inform people the magnitude of the load. If an exact amount of mass is not available, the approximately masses (range of the masses) of the loads should still be provided, so that people can estimate the magnitude of the loads before lifting.

Lifting over a constraining barrier has been shown to increase the risk of the lower back injury. The previous study (McKean and Potvin, 2001) found that when lifting over a constraining barrier subjects tended to move the load further from the lower back in the horizontal direction. They also found that subjects required more trunk flexion and knee extension when lifting over the constraining barrier. McKean and Potvin (2001) did not estimate the stresses (i.e., moments, compressive forces) at the lower back and only compared lifting tasks between a single constraining barrier to the non-constraining barrier. They used the changes of muscle activity (EMG) in the thoracic and lumbar regions to represent the spinal loading when lifting over the constraining barrier. The contribution of this research was to estimate the peak compressive forces at L5/S1 when lifted over the constrained condition of five different heights of constraining barrier. The results showed that when subjects lifting over the constrained conditions, the peak compressive forces in the lower back significantly increased compared to the non-constrained condition. This was because subjects moved the load further from the lower back when lifting during the constrained conditions. When lifting during the constrained conditions, subjects increased the sacral angle, but decreased the knee flexion. In the current research, the peak compressive forces at L5/S1 showed a significantly quadratic trend. However, the magnitude of the difference of the peak compressive forces was small. In order to prevent the risks of the lower back injury, people should avoid lifting a load over a barrier if possible. This is because the constrained condition significantly increases horizontal displacement between the load and the lower back, which will produce greater stresses on the lower back.

Internal and external environment of variables should be considered when assessing the potential risks of back injury. The components that cause changes to the peak compressive forces should be analyzed, some components may decrease peak compressive forces but increase other loads. Individuals respond to fatigue in different ways.

Future studies

Study of fatigue as the result of different protocols while lifting should be considered as a future research, instead of just comparing fatigue conditions, pre and post fatigue conditions for a specific protocol.

The effect of unexpected loads with different masses and how individuals adapt it to the subsequent lifts probably could be considered as a future research.

Future research and more data collection of lifting tasks under different levels of constraining barrier are required to emphasize the meaning of the quadratic trend of the peak compressive forces at L5/S1. The data collection of lifting over the constrained condition, where the height of constraining barrier is between 20% and 80% or higher than 140% of knee height, is probably interesting for future investigation.

APPENDIX A: INFORMED CONSENT

Appendix A contains the information for the Informed Consent to participate in research form. There are three different forms of the Informed Consent to participate in research form. The Iowa State University Human Subjects Review Committee prior to the lifting experiments in this study approved these forms. The Informed Consent form was reviewed and signed by the subject in this study after any questions about the research were addressed prior to the experiments.

Informed Consent to Participate in Research

Department of Industrial and Manufacturing System Engineering
Iowa State University
Ames, IA 50011

You are being asked to volunteer as a participant in a research study. This form is designed to provide you with information about this study and to answer your questions.

1. Title of Research Study

Influence of fatigue on L5/S1 Compressive Force During Manual Lifting

2. Project Directors

Name:	Iwan Budihardjo	Name:	Tim R Derrick
Address:	3013 Black Engineering	Address:	249 Forker Building
Email:	ibudihar@iastate.edu	Phone:	294 8438
		Email:	tderrick@iastate.edu

3. Purpose of the Research

The purpose of this study is to examine the loads on the lower back while subjects lift a crate until fatigued.

4. Procedures for this Research

Orientation for Lifting Trials

Prior to the start of the study, there will be an orientation session to familiarize you with the lifting motions to be analyzed and the equipment used during the data collection. At this time you will be able to ask questions and obtain further information about all aspects of the study. You will be asked to perform lifting trials until you feel fatigued. You will be able to use the lifting technique that you feel most comfortable with. The object to be lifted is a crate containing 10 kilograms (22 pounds) of weights. The motion involves lifting the crate from ground level in front of the body to about the knuckle height. The crate will then be returned to the ground and the next lift will be initiated. You will perform 10 lifts per minute for up to 1 hour or until you become fatigued. Every a minutes you will be shown a scale that ranges from 0 (no exertion) to 10 (maximal exertion). You will point out the number that most closely approximates your state of fatigue. The lifts will be stopped if you reach 9 or 10 on the scale.

Anthropometric Measurements, Marker and Skin Surface Electrode Placements

During the lifting trials, you will be asked to wear black biking shorts and tank top provided by the researchers. This clothing will make it more accurate for the computer to locate your joints. Before performing the lifting trials, a set of anthropometric measurements will be taken to determine the size and shape of your body segments. These measurements will be taken with a standard scale, a measuring tape, and a beam caliper. A set of 1-inch diameter reflective balls will be attached to highlight anatomical landmarks on your body. These markers will be attached to your skin or clothing by way of double-sided adhesive tape. Pairs of disposable surface electrodes will be attached on either side of your spine on the lower back. The electrode sites will be cleansed with alcohol swabs before the electrodes are attached. These electrodes will record the muscle activity in the muscles of the lower back.

Data Collection

During the lifting trials, the markers that have been placed on your body will be tracked by four video cameras. An additional conventional video camera will also record you to aid in marker identification during data analysis. Your digital image will be stored on the computer and deleted at the conclusion of the study. The skin surface electrodes that have been placed on your body will be used to record muscle activity so the fatigue can be determined during the lifting trials. You will be standing with the right foot on a device that will record the forces while performing the lifting trials. Tracking markers by video cameras and force platform measurements are common procedures used in biomechanics. The total time for orientation, measurement, and the lifting trial is expected to be about two hours.

5. Potential Risks or Discomforts

Some minor irritation might occur upon removal of the markers or electrodes. Some muscle soreness may result from the repeated lifts. There are no invasive procedures used in this study.

6. Potential Benefits to you or others

There will be no direct benefits to you as subject in this study. This study may lead to a better method of determining joint forces, moments and compressive forces on the lower back during lifting. It may also lead to improved strategies to deal with the effects of fatigue while lifting.

7. Alternate Treatment or Procedures, if applicable

You cannot participate in this study if you have any history problems with your back. You have the option of not participating in the study. You are also free to withdraw from the study at any time without consequence.

Please circle your answer:

Do you have any history of back problems? **YES** **NO**

Do you understand that you will not receive money for your participation in this study? **YES** **NO**

Do you understand that you are free to withdraw your consent and discontinue participation in this research project at any time without prejudice? **YES** **NO**

Emergency treatment of any injuries that may occur as a direct result of participation in this research will be treated at the Iowa State University Student Health Services. Student Services Building, and/or referred to Mary Greely Medical center or another physician. Compensation for treatment of any injuries that may occur as a direct result of participation in this research may or may not be paid by Iowa State University depending on the Iowa Tort Claims Act. Claims for compensation will be handled by the Iowa State University Vice president for Business and Finance.

Your questions on any aspect of this research project are welcomed. At the conclusion of this study you may request a summary of the results. Your individual results will be kept confidential and should the data be used in a publication of the results, your name or any identifying characteristics will not be reported.

Signatures

I have fully explained to --
-------the nature and purpose of the above study and the benefits and risks that are involved in participation of the study. I have answered and will answer all questions to the best of my ability.

---	---
Signature of Principal Investigator Obtaining Consent	Date

I have been fully informed of the above-described procedure with its possible benefits and risks and I have received a copy of this description. I have given permission for my participation in this study.

---	---
Signature of Participant	Date
---	---
Signature of Witness	Date

Informed Consent to Participate in Research

Department of Industrial and Manufacturing System Engineering
Iowa State University
Ames. IA 50011

You are being asked to volunteer as a participant in a research study. This form is designed to provide you with information about this study and to answer your questions.

1. Title of Research Study

Influence of the knowledge of load magnitude on L5/S1 compressive force during manual lifting

2. Project Directors

Name: Iwan Budihardjo
Address: 3013 Black Engineering
Email: ibudihar@iastate.edu

Name: Tim R Derrick
Address: 249 Forker Building
Phone: 294 8438
Email: tderrick@iastate.edu

3. Purpose of the Research

The purpose of this study is to examine how subjects adapt to the loads on the lower back while they lift boxes with and without knowledge of load magnitude.

4. Procedures for this Research

Orientation for Lifting Trials

Prior to the start of the study. there will be an orientation session to familiarize you with the lifting motions to be analyzed and the equipment used during the data collection. At this time you will be able to ask questions and obtain further information about all aspects of the study. You will be asked to perform 8-9 series of 5 lifting trials. The mass of the load will be either low (3 kg, 6.6 pounds) or high (17 kg, 37.4 pounds). Sometimes you will be told which mass you are lifting and sometimes you will not. During each series of lifts you will initiate a lift every 6 seconds. You will be able to use the lifting technique that you feel most comfortable with. The motion involves lifting the box from ground level in front of the body to about the knuckle height. The box will then be returned to the ground and the next lift will be initiated. You can have as much rest between each series of lifts as needed to recover.

Anthropometric Measurements and Marker Placements

During the lifting trials, you will be asked to wear black biking shorts and tank top provided by the researchers. This clothing will make it more accurate for the computer to locate your joints. Before performing the lifting trials, a set of anthropometric measurements will be taken to determine the size and shape of your body segments. These measurements will be taken with a standard scale, a measuring tape, and a beam caliper. A set of 1-inch diameter reflective balls will be attached to highlight anatomical landmarks on your body. These markers will be attached to your skin or clothing by way of double-sided adhesive tape.

Data Collection

During the lifting trials, the markers that have been placed on your body will be tracked by four video cameras. An additional conventional video camera will also record you to aid in marker identification during data analysis. Your digital image will be stored on the computer and deleted at the conclusion of the study. You will be standing with the right foot on a device that will record the forces while performing the lifting trials. Tracking markers by video cameras and force platform measurements are common procedures used in biomechanics. The total time for orientation, measurement, and the lifting trial is expected to be about two hours.

5. Potential Risks or Discomforts

Some minor irritation might occur upon removal of the markers. Some muscle soreness may result from the repeated lifts. There are no invasive procedures used in this study.

6. Potential Benefits to you or others

There will be no direct benefits to you as subject in this study. This study may lead to a better method of determining joint forces, moments and compressive forces on the lower back during lifting. It may also lead to improve lifting strategies.

7. Alternate Treatment or Procedures, if applicable

You cannot participate in this study if you have any history problems with your back. You have the option of not participating in the study. You are also free to withdraw from the study at any time without consequence.

Please circle your answer:

Do you have any history of back problems? **YES** **NO**

Do you understand that you will not receive money for your participation in this study? **YES** **NO**

Do you understand that you are free to withdraw your consent and discontinue participation in this research project at any time without prejudice? **YES** **NO**

Emergency treatment of any injuries that may occur as a direct result of participation in this research is available at the Iowa State University Thomas B. Thielen Student Health Center, and/or referred to Mary Greeley Medical Center or another physician or medical facility at the location of the research activity. Compensation for any injuries will be paid if it is determined under the Iowa Tort Claims Act, Chapter 669 Iowa Code. Claims for compensation should be submitted on approved forms to the State Appeals Board and are available from the Iowa State University Office of Risk Management and Insurance.

Your questions on any aspect of this research project are welcomed. At the conclusion of this study you may request a summary of the results. Your individual results will be kept confidential and should the data be used in a publication of the results, your name or any identifying characteristics will not be reported.

Signatures

I have fully explained to ---
-------the nature and purpose of the above study and the benefits and risks that are involved in participation of the study. I have answered and will answer all questions to the best of my ability.

--- ---

Signature of Principal Investigator Obtaining Consent

Date

I have been fully informed of the above-described procedure with its possible benefits and risks and I have received a copy of this description. I have given permission for my participation in this study.

--- ---

Signature of Participant

Date

--- ---

Signature of Witness

Date

Informed Consent to Participate in Research

Department of Industrial and Manufacturing System Engineering
Iowa State University
Ames, IA 50011

You are being asked to volunteer as a participant in a research study. This form is designed to provide you with information about this study and to answer your questions.

1. Title of Research Study

The influence of barrier height on L5/S1 compressive force during manual lifting

2. Project Directors

Name: Iwan Budihardjo
Address: 3013 Black Engineering
Email: ibudihar@iastate.edu

Name: Tim R Derrick
Address: 249 Forker Building
Phone: 294 8438
Email: tderrick@iastate.edu

3. Purpose of the Research

The purpose of this study is to examine how subjects adapt to lifting a load over barriers of various heights.

4. Procedures for this Research

Orientation for Lifting Trials

Prior to the start of the study, there will be an orientation session to familiarize you with the lifting motions to be analyzed and the equipment used during the data collection. At this time you will be able to ask questions and obtain further information about all aspects of the study. You will be able to use the lifting technique that you feel most comfortable with. The object to be lifted is a crate containing 10.3 kilograms (22 pounds) of weights. A simulated barrier with different heights will be placed between you and the crate. There are 5 lifting conditions that represent the heights of the barrier. They are 0% (no barrier), 80%, 100%, 120% and 140% of knee height as the height of barrier conditions. The motion involves lifting the crate from ground level in front of the body to a shelf at knuckle height. The crate will then be returned to the ground and the next lift will be initiated. You will perform 10 lifts during a one minute interval for each condition. The lifting trials will be assigned randomly for each subject.

Anthropometric Measurements and Marker Placement

During the lifting trials, you will be asked to wear black biking shorts and tank top provided by the researchers. This clothing will make it more accurate for the computer to locate your joints. Before performing the lifting trials, a set of anthropometric measurements will be taken to determine the size and shape of your body segments. These measurements will be taken with a standard scale, a measuring tape, and a beam caliper. A set of 1-inch diameter reflective balls will be attached to highlight anatomical landmarks on your body. These markers will be attached to your skin or clothing by way of double-sided adhesive tape.

Data Collection

During the lifting trials, the markers that have been placed on your body will be tracked by four video cameras. An additional conventional video camera will also record you to aid in marker identification during data analysis. A digital image of the reflective markers will be stored on the computer and deleted at the conclusion of the study. You will be standing with the right foot on a device that will record the forces while performing the lifting trials. Tracking markers by video cameras and force platform measurements are common procedures used in biomechanics. The total time for orientation, measurement, and the lifting trial is expected to be less than two hours.

5. Potential Risks or Discomforts

Some minor irritation might occur upon removal of the markers. Some muscle soreness may result from the repeated lifts.

6. Potential Benefits to you or others

There will be no direct benefits to you as subject in this study. This study may lead to a better method of determining joint forces, moments and compressive forces on the lower back during lifting. It may also lead to improved strategies to deal with the effects of different heights of simulated barrier while lifting.

7. Alternate Treatment or Procedures, if applicable

You cannot participate in this study if you have any history problems with your back. You have the option of not participating in the study. You are also free to withdraw from the study at any time without consequence.

Please circle your answer:

Do you have any history of back problems? **YES** **NO**

Do you understand that you will not receive money for your participation in this study? **YES** **NO**

Do you understand that you are free to withdraw your consent and discontinue participation in this research project at any time without prejudice? **YES NO**

Emergency treatment of any injuries that may occur as a direct result of participation in this research is available at the Iowa State University Thomas B. Thielen Student Health Center, and/or referred to Mary Greeley Medical Center or another physician or medical facility at the location of the research activity. Compensation for any injuries will be paid if it is determined under the Iowa Tort Claims Act, Chapter 669 Iowa Code. Claims for compensation should be submitted on approved forms to the State Appeals Board and are available from the Iowa State University Office of Risk Management and Insurance.

Your questions on any aspect of this research project are welcomed. At the conclusion of this study you may request a summary of the results. Your individual results will be kept confidential and should the data be used in a publication of the results, your name or any identifying characteristics will not be reported.

Signatures

I have fully explained to --
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--	--
Signature of Principal Investigator Obtaining Consent	Date

I have been fully informed of the above-described procedure with its possible benefits and risks and I have received a copy of this description. I have given permission for my participation in this study.

--	--
Signature of Participant	Date

--	--
Signature of Witness	Date

APPENDIX B: ANTHROPOMETRIC TABLE

Adjusted Body Segment Parameter Data for males (Zatsiorsky et al. 1990. adjusted by de Leva. personal communication. April 14. 1995)

Segment	Proximal Reference	Distal Reference	Relative Mass	CM Location (from proximal Reference)	Radius of gyration (ML direction)
Foot	heel	toe	0.0137	0.4415	0.257
Shank/leg	knee joint center	ankle joint center	0.0433	0.4459	0.255
Thigh	hip joint center	knee joint center	0.1416	0.4095	0.329

APPENDIX C: THE L5/S1 AND PELVIS MODELS

Calculation of the L5/S1 global coordinates

The three-dimensional L5/S1 global coordinate when subject was in dynamic trials (P_G) was obtained by using the transformation matrix (T) from the local coordinate frame L. to the global coordinate frame G. as follow:

$$P_G = L_G + [R]\,P_L$$
$$= [T]P_L$$

The transformation is given in terms of the vector L_G, which represents the location of the origin of the local frame relative to the global one, and the matrix [R], representing the orientation of the local frame. And, by using a (4 3 4)-transformation matrix [T], translation and rotation can be simultaneously represented by one matrix multiplication, as follows:

$$[T] = \begin{bmatrix} 1 & : & 0 \quad 0 \quad 0 \\ \cdots & \cdots & \cdots\;\;\cdots\;\;\cdots \\ [L] & : & [R] \end{bmatrix}$$

The elements of the matrix $[L_G]$, the location of the origin of the local frame relative to the global were the three-dimensional orientation of the hip joint center. And, the elements of the matrix [R] were three vector columns that represented by the three-dimensional unit vector orientations of the three markers in the pelvis segment.

$$[L_G] = \begin{bmatrix} L_X \\ L_Y \\ L_Z \end{bmatrix} \qquad [R] = \begin{bmatrix} R_{X1} & R_{X2} & R_{X3} \\ R_{Y1} & R_{Y2} & R_{Y3} \\ R_{Z1} & R_{Z2} & R_{Z3} \end{bmatrix}$$

The first column of matrix [R] was the three-dimensional unit vector from hip to sacrum orientations. Then, the second column of matrix [R] was the three-dimensional unit vector of the cross product between the vector from hip to iliac crest orientations and the unit vector of the first column of matrix [R]. And finally, the third column of matrix [R] was the three-dimensional unit vector of the cross product between the vector from hip to sacrum orientation and the unit vector of the second column of matrix [R].

Matrix $[P_L]$ was the three-dimensional L5/S1 local coordinate when the subject was in standing erect. This local coordinate was obtained by using the inverse transform matrix formula from the global system to the local one, when the subject was in standing erect position:

$$P_L = (\text{transpose } [R])\,(P_G - L_G)$$

$$P_L = (\text{inverse}[T])\,P_G$$

The inverse of transformation matrix [T] is

$$\begin{bmatrix} 1 & : & 0 \quad 0 \quad 0 \\ \ldots & \ldots & \ldots \quad \ldots \quad \ldots \\ -\,\text{trans}[R]\,[L] & : & \text{trans}[R] \end{bmatrix}$$

The element of matrix $[P_G]$ was the three-dimensional L5/S1 joint orientation coordinate when the subject was standing erect (19.5% from hip between hip and shoulder joint centers). The elements of matrix $[L_G]$ and [R] were the same as described above.

The three-dimensional L5/S1 local coordinates when subject was standing erect were stored as subject's L5/S1 local coordinates. It was used by the transformation matrix formula to obtain the three-dimensional L5/S1 global coordinate (P_G) when subject was in dynamic trials.

Elliptical solid

To obtain the center of mass (CM) of pelvis in longitudinal position (Z) relative to the pelvis length (L), the following calculation is required:

$$Z = [(G_{21}(A, B) / G_{20}(A, B))\,L] / L\,,$$

Which G_{21} (A, B) and G_{20}(A, B) are the functions that derived from anthropometric measurements. The CM of pelvis is relative to the pelvis length to the hip joint center as the proximal point.

The moment of inertia of pelvis to the center of mass on the ML axis, I_{YY}, is obtained by:

$$I_{YY} = I_Y - Z^2.$$

Which Z is the CM of pelvis in longitudinal position, and I_Y is the moment of inertia of pelvis to the proximal point on the ML axis. I_Y is obtained by:

$$I_Y = (\Pi/4)\,\rho\, L\, G_{40}(A, A, A, B) + \Pi\, \rho\, (L\hat{}3)\, G_{22}(A, B).$$

Which G_{40}(A, A, A, B) and G_{22}(A, B) are the functions that derived from anthropometric measurements. ρ is uniform density of pelvis (Pearsall, et al., 1996).

G_{20}(A, B), G_{21} (A, B), G_{22}(A, B), and G_{40}(A, A, A, B) are the functions that derived from the anthropometric measurements, the radius of the hip and L5/S in medial lateral (ML) and anterior posterior (AP) directions.

$$G_{20}(A, B) = F_{22}(A, B)/3 + F_{21}(A, B)/2 + F_{20}(A, B).$$

$$G_{21}(A, B) = F_{22}(A, B)/4 + F_{21}(A, B)/3 + F_{20}(A, B)/2.$$

$$G_{22}(A, B) = F_{22}(A, B)/5 + F_{21}(A, B)/4 + F_{20}(A, B)/3.$$

$$G_{40}(A, A, A, B) = F_{44}(A, A, A, B)/5 + F_{43}(A, A, A, B)/4 + F_{42}(A, A, A, B)/3 + F_{41}(A, A, A, B)/2 + F_{40}(A, A, A, B).$$

$F_{20}(A, B) = A_0B_0$

$F_{21}(A, B) = A_0(B_1 - B_0) + (A_1 - A_0)B_0$

$F_{22}(A, B) = (A_1 - A_0)(B_1 - B_0)$

$F_{40}(A, A, A, B) = A_0A_0A_0B_0.$

$F_{41}(A, A, A, B) = (A_1 - A_0)\, A_0A_0B_0 + A_0(A_1 - A_0)\, A_0B_0 + A_0A_0(A_1 - A_0)B_0 + A_0A_0A_0(B_1 - B_0).$

$F_{42}(A, A, A, B) = A_0A_0(A_1 - A_0)(B_1 - B_0) + A_0(A_1 - A_0)A_0(B_1 - B_0) + A_0(A_1 - A_0)(A_1 - A_0)B_0 + (A_1 - A_0)A_0A_0(B_1 - B_0) + (A_1 - A_0)A_0(A_1 - A_0)B_0 + (A_1 - A_0)(A_1 - A_0)A_0B_0$

$F_{13}(A, A, A, B) = A_0(A_1 - A_0)(A_1 - A_0)(B_1 - B_0) + (A_1 - A_0)A_0(A_1 - A_0)(B_1 - B_0) + (A_1 - A_0)(A_1 - A_0)A_0(B_1 - B_0) + (A_1 - A_0)(A_1 - A_0)(A_1 - A_0)B_0.$

$F_{14}(A, A, A, B) = (A_1 - A_0)(A_1 - A_0)(A_1 - A_0)(B_1 - B_0).$

Where A_1 is the radius of L5/S1 in AP direction, A_0 is the radius of hip in AP direction, B_1 is the radius of L5/S1 in ML direction, and B_0 is the radius of hip in ML direction.

APPENDIX D: MOTOR UNIT AND MUSCLE FIBER TYPES

The motor neuron is the functional unit of the nervous system carrying information to and from the nervous system. A motor unit is composed of a motor neuron and all of the muscle fibers it innervates. It is the smallest functional unit of muscular shortening or force production. Each muscle has many motor units. The number of fibers in a motor unit is dependent on the precision of movement required of that muscle. The average number of fibers per neuron is somewhere between 100 and 200 muscle fibers (Enoka. 1988. Basmajian. 1978).

There are three different types of motor units, corresponding to the three fiber types: slow-twitch oxidative (type I), fast-twitch oxidative (type IIa), and fast-twitch glycolytic (type IIb). All of the muscle fibers in a motor unit are of the same type. Within a muscle there is a mixture of fiber types. These fibers and motor unit types are basically genetically determined, but they may change with training. The fiber type characteristics are as follows:

- Slow-twitch oxidative

It is small size and slow shortening speed, so the force production is low. It is low in anaerobic capacity, but very high in aerobic capacity and then low fatigability (fatigue resistant).

- Fast-twitch oxidative

It is large size and fast shortening speed, then the force production is high. The capacities of aerobic and anaerobic are medium, so this type is medium resistant of fatigue.

- Fast-twitch glycolytic

It is the largest size and fastest speed, and the highest in force production. The anaerobic capacity is very high, but the aerobic is very low, so this type is high fatigability (non-fatigue resistant).

The excitation of a motor unit is an all-or-nothing principle. To increase tension can be accomplished by:

- Increasing the number of stimulated motor units (recruitment)

Each motor unit has a stimulation threshold at which it will begin to produce force. Small motor units have a lower threshold than large motor unit, therefore they are recruited first (size principle). Recruitment is ordered: type I recruited first (lowest threshold), then type IIa recruited second, and type IIb recruited last (highest threshold).

- Increasing the stimulation rate of the active motor units (rate coding).

Rate of coding or frequency of coding is the frequency of motor unit firing that influence the amount of force or tension developed by the muscle. The rate of coding also varies with the fiber type and change with the type of movement. In small muscle types, all of the motor units are usually recruited and activated when the external force of the muscle is at levels of 30-50% of the maximum voluntary contraction level (Enoka, 1988). In the large muscle types, there is recruitment of motor units all through the total force range so that muscles are still recruiting more motor units at 100% maximum voluntary contraction (Enoka, 1988).

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