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EFFECT OF LOAD ASYMMETRY ON INTERNAL LOADING OF THE TRUNK

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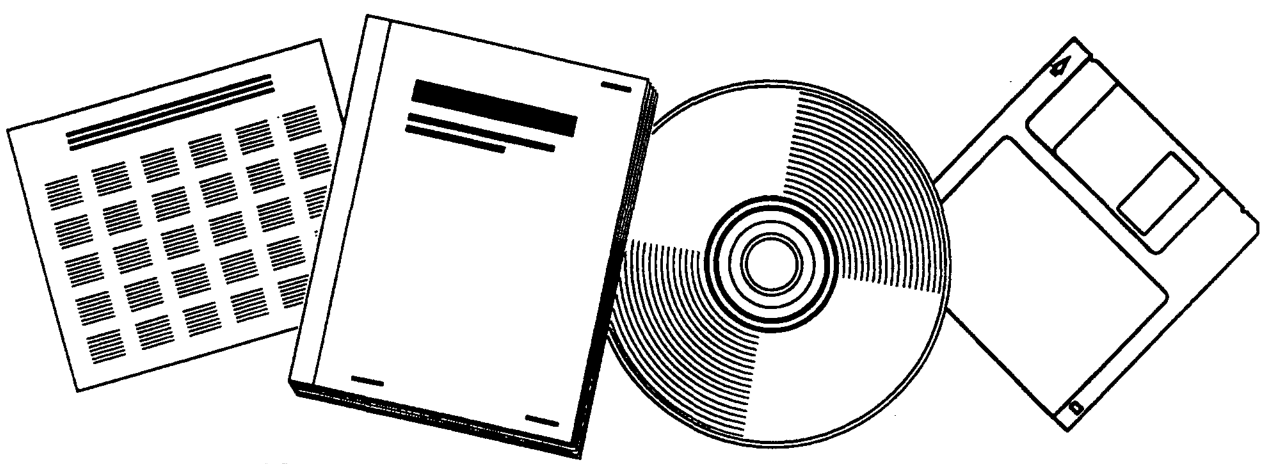

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Effect of Load Asymmetry on Internal Loading of The Trunk

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16. Abstract (Limit: 200 words)

The effects of changes in the magnitude and angle of the net resultant force on the electromyographic (EMG) activity of trunk muscles, and the forces acting on the lumbar spine during upright asymmetric isometric trunk exertion were evaluated. The activities of ten trunk muscles were quantified using wire and surface EMG during maximal and submaximal isometric exertion. Subjects included 15 healthy volunteers, mean age 29.8 years. Muscle parameters were calculated from computed tomography images. The normalized root mean square (NRMS) EMG of the trunk muscles were affected by trunk movement magnitude and angle. An EMG driven model of lumbar spine load was used to calculate compression and shear forces. The author concludes that the selected trunk muscles were significantly affected by the magnitude and direction of the net resultant moment. The NRMS-EMG activity of the trunk muscles was higher during combined (asymmetric) isometric exertion compared to pure (symmetric) isometric trunk exertion. The controllability and capability of high magnitude trunk moments were significantly affected by the direction of the net resultant moment. The results may have special applications for ergonomic job evaluation and job design. The author recommends that repetitive tasks involving asymmetric trunk postures should be avoided as should the combination of flexion and twisting of the trunk.

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Significant Findings

Asymmetrical trunk postures are common in industry. Some evidence from the epidemiological literature suggests that these postures are associated with higher frequency of reported injuries to the lower back. This study was undertaken to increase the knowledge of the loads on the lumbar spine during isometric asymmetric trunk exertions.

Electromyographic (EMG) activity of trunk muscles and load on the lumbar spine during dynamic asymmetrical exertion has been studied previously. High spinal forces reported by these studies is partially due to a change of trunk posture and by that changes in trunk muscle alignments, and partially due to changes in net internal resultant moments. The aim of this study was to investigate the effect of changes in the magnitude and angle of the net resultant force on the EMG activity of trunk muscles and thereby the forces acting on the lumbar spine during upright asymmetric isometric trunk exertion.

The result of the present study demonstrated the effect of magnitude and direction of net resultant moments on the normalized root mean square (NRMS) EMG activity of ten selected trunk muscles. Three main conclusions were drawn from this study:

1. The ten selected trunk muscles were significantly affected by the magnitude and direction of the net resultant moment.
2. The NRMS-EMG activity of the trunk muscles was higher during combined (asymmetric) isometric exertion compared to pure (symmetric) isometric trunk exertion.

3. The controllability and capability of high magnitude trunk moments were significantly affected by the direction of the net resultant moment.

This study used an EMG driven model to estimate loads on the lumbar spine at L3-L4 level. Previous studies in the literature suggest that EMG-driven models should be used for estimation of spinal forces and risk of injury during asymmetrical exertion. The present study demonstrates the shortcoming of these models and suggestions for enhancement of these models. The results also show that EMG-driven models are sensitive to the assumptions made with respect to EMG-force relationship of trunk muscles. The trunk moments and spinal forces based on three selected assumptions are presented. The results of the statistical analysis suggest that the sagittal moment, the lateral shear force, and the compression force on the lumbar spine were highly and significantly ($p<0.03$) affected by the mathematical formulation of force in these models.

Two sites have been suggested in the literature for monitoring the myoelectrical activity of the internal oblique (IOB) muscles, the anterior triangle and the posterior triangle. Placement of surface electrodes in previous studies at the posterior triangle area suggests that the IOB muscle be a trunk extensor. This finding is opposite to the studies that monitored the EMG activity of the IOB muscle at the anterior triangle area. In the present study, wire electrodes were used to investigate the functional behavior of the IOB muscle during pure and combined trunk exertion. The result of this study shows that the IOB muscle is a trunk flexor both in symmetrical and asymmetrical isometric trunk exertions. The potential problems and error due to use of the lumbar posterior triangle for monitoring EMG activity of the IOB muscle have been discussed based on results of individual CT-scan images of the participating subjects in this study and dissection of one cadaver. This study demonstrated that the EMG-driven models are sensitive to assumptions made with respect to the geometrical description of the trunk muscles. The results of the statistical analysis suggest that the sagittal moment, shear forces, and muscle gains are significantly ($p<0.002$) effected by these assumptions.

Usefulness of Findings

The results of the present study have both practical and theoretical implications for studying low back pain and risk of injury during asymmetrical trunk exertion. The results have special applications for ergonomic job evaluation and job design. It was demonstrated that controllability and capability of trunk exertion decreases as the task becomes asymmetrical. Therefor job design and especially repetitive tasks involving asymmetrical trunk postures should be avoided. This study shows that increased asymmetric postures involving flexion and rotation of the trunk significantly reduce the exertion level and the controllability of exertion. The findings also indicate a significant increased load on the lumbar spine. Therefor in job design the combination of flexion and twisting of the trunk should be avoided to the extent possible. This information is particular important for new design and redesign of a job station and product development. Industrial designers, employers and unions should be aware of these findings in an attempt to lower the risk for low back injuries.

The capability of symmetrical trunk exertion cannot be used as an indication for the capacity of the individual during asymmetrical trunk exertion. This study indicates that testing an individual in a symmetrical posture will not predict the individual's capacity in an asymmetric trunk exertion task if trunk flexion and twisting must be performed simultaneously on the job.

The study also has implications for the clinician interested in prevention and rehabilitation and for the ergonomists who are involved in occupational rehabilitation (work conditioning and work hardening programs), and functional capacity evaluation of an individual with an injury to the lumbar spine. Rehabilitation programs should design work out protocol including increased endurance, strength and coordination exercises to increase the controllability and stamina in flexion combined with twisting of the trunk to enhance the individual's personal prevention of low back injury. This protocol could also be used as programs in industry for primary prevention of low back injury.

In research, the electromyography (EMG) driven models have been extensively used in the literature to investigate the effect of symmetrical and asymmetrical trunk exertion and the effect on spinal loads. The present study demonstrates that the outcomes of EMG-driven models are sensitive to the anatomical and physiological assumption made in the mathematical formulation of these models. These assumptions significantly affect the estimated spinal forces and moments. Since forces currently cannot be measured directly on the spinal structures, there is no method for measuring the validity of one model compared with other models. The interpretation of the results of different models should be based on: 1) the validity of assumptions made in the formulation of forces in these models, 2) the accuracy of predicted trunk moments, and 3) the range of spinal force estimated by a model. The EMG-driven models are the best tools for investigating the effect of industrial activities on spinal forces and risk of injury. The refinement of these models enhances our understanding of human performance and individual limitations. The present study provides a new method of formulating muscle force in EMG-driven models. Advantages of the new method are: 1) individual muscle gain can be calculated, 2) the formulation of model is flexible to the assumption made with respect to EMG-force relationship of the trunk muscles.

EMG-driven models provide a tool for investigating the effect of industrial and clinical guidelines based on physical condition of the patient. Based on the result of these models, the efficacy of guidelines may be better evaluated in the future.

Abstract

In industry, carrying and lifting objects asymmetrically is normal daily practice. This situation is hazardous to the musculoskeletal system due to an increase in coactivation of the musculature and an increase in the load on the spine.

The aim of this study is to quantify the activities of ten trunk muscles by using wire and surface electromyography (EMG) during maximal and submaximal isometric exertion, under pure and combined loading conditions of the trunk. Combined loading is defined as the vectorial sum of moments in the sagittal and transverse planes. These planes are selected based on their prevalence in industrial tasks and low back injuries.

To calculate the effect of internal loading, the muscle parameters are calculated from CT-scan images of each participant. Individual muscle forces are estimated using an EMG-driven model. The compression and shear forces are calculated under suggested conditions. The effects of pure exertion (planar exertion) and combined exertion on the patterns of trunk muscle recruitment and the compression and shear forces are calculated.

The primary hypotheses of this study are:

1. The mean of NRMS-EMG of the ten selected trunk muscles will change significantly with the orientation and magnitude of the net resultant moments of the trunk.
2. The result of the EMG-driven models will be significantly affected by assumptions made with respect to muscle EMG-force relationship.

3. The result of the EMG-driven models will be significantly affected by assumptions made with respect to geometrical values and function of trunk muscles.

To test the first hypothesis, fifteen healthy volunteers with a mean age of 29.8 (6.4) years and height 168.2 (8.7) cm and weight 69.9 (12.9) Kg were tested. The results of MANOVA indicated that the NRMS-EMG of ten selected trunk muscles were significantly ($P<0.0001$) affected by the magnitude and angles of the net resultant trunk moments. The muscles showed higher activity during combined exertion compared to pure exertion. At higher planar exertion, the capability and controllability of the trunk muscles were significantly reduced during combined exertions.

To test the second and third hypotheses, compression and shear forces for six subjects were calculated based on the EMG-driven models of lumbar spine load. The statistical analysis suggests that the sagittal moment, lateral shear force, and compression force estimated by the EMG-driven models are significantly ($P<0.001$) affected by the assumptions made with respect to the EMG-force relation of individual trunk muscles. For the third hypothesis, the statistical analysis demonstrates that sagittal moments, shear forces, and estimated gains are significantly affected by geometrical parameters and the assumptions made with respect to the role of the internal oblique muscles.

A new method of calculating muscle force has been suggested in this study. The suggested method allows for an estimation of gain for each individual muscles and group of muscles according to function.

The present study will provide a theoretical and methodological refinement of EMG-driven models. The information presented in this study enhances our understanding of mechanical and neuromuscular performance in humans and the risk of low back injury under asymmetrical trunk exertions.

Alterations to the Original Proposed Project

The main focus of this study was to collect sets of data that are reliable for testing our hypotheses and thereby to improve the biomechanical load model of the lumbar spine. Therefore, the electrode placements, reliability, and validity of the collected EMG signal were a major concern in the early stage of study.

Our initial attempt in data collection showed poor validity for the internal oblique (IOB) muscle EMG signal. This led to a detailed investigation of the electrode placement of the muscle. Two sites have been suggested in the literature for monitoring the EMG activity of the IOB muscle, the anterior and the posterior or lumbar triangle (see Chapter 4 for definition and detailed discussion). Original peer reviewed, published studies were used as a base for our first study design. None of these studies reported the validity of the electrode placement in the lumbar triangle. No investigator reported any of our concerns with respect to electrode placement (i.e., size of the lumbar triangle with reference to small and large size individuals, margins of error for electrode placement, and anatomical individual variations).

The main question in the early stage of our study was the margin of error in placement of the electrode in the lumbar triangle. This led us to inspect the size of lumbar triangle with respect to size electrodes in a dissection study of one cadaver. Detail discussion related to this issue and our findings are reported in chapter 4. The result of the dissection study led us to believe that the validity of the IOB muscle electrode placement should be further studied and wire electrodes should be used to monitor the myoelectrical activity of the IOB muscle. This decision was made based on following facts:

- Eigthy percent of the tasks in this study involved maximal or submaximal trunk exertion in transverse plane.
- The IOB muscle has a major role in transverse trunk exertion.
- The lumbar triangle size is too small for placement of two surface electrodes.
- Misplacement of surface electrodes may magnify the EMG signal collected from the IOB muscle and consequently jeopardize the results of our hypothesis testing and the calculation of lumbar spinal forces.
- Inspection of the individual CT-scan images of the first tested subjects showed that the posterior end of the IOB muscle does not form the lumbar triangle as it is explained in general anatomical text books (see Chapter 4 for more details).

The original testing protocol was modified to include placement of wire electrodes in the IOB muscles under CT-scan imaging. Due to technical limitations associated with wire electrode technique, our sample size was reduced to six subjects for the second phase of the study. More subjects were needed for statistical analysis for the hypothesis related to compression and shear forces under exertion testing conditions. Therefore, the last hypothesis in the original study was not tested.

During the course of this study, we discovered theoretical problems in formulation of the forces in the EMG-driven models. Chapters 3 and 4 are addressing some of these problems.

A new method of calculation of muscle forces has been suggested to overcome these problems. This was not part of originally proposed study.

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LIST OF ABBREVIATIONS

A/D	Analog to Digital
ANOVA	Analysis of Variance
CT	Computed Tomography
CNS	Central Nervous System
EMG	Electromyographic Signal
EOB	External Oblique Muscles
ES	Erector Spinae muscles
CI	Confidence Interval
IAP	Intra-Abdominal Pressure
IOB	Internal Oblique Muscles
LAT	Latissimus Dorsi Muscles
LBP	Low Back Pain
MANOVA	Multivariate Analysis of Variance
MRI	Magnetic Resonance Imaging
MVE	Maximum Voluntary Exertion
NIOSH	National Institute for Occupational Safety and Health
NRMS-EMG	Normalized Root Mean Square of Electromyographic Signal
NS	Not Significant
OR	Odd Ratio
RA	Rectus Abdominus
VDT	Video/Visual Display Terminal

Chapter I

Introduction

1.1 Review of the Literature

Despite increased automation in the workplace (Marras et al., 1991) and modern advances in medical diagnosis, prevention, and treatment, low back pain (LBP) continues to be widespread amongst industrial workers. Most researchers in the past two decades attempted to verify the risk factors of occupational low back pain and to reduce the effect of these factors in common industrial tasks.

The epidemiological studies have identified work intensity, static posture, frequent bending and twisting, lifting, pushing and pulling, and repetition as occupational risk factors associated with LBP (Genaidy et al., 1994; Corlettt and Richardson 1981; Andersson 1985; Hagberg et al., 1995; Marras et al., 1993; Punnett et al., 1991). These risk factors were statistically or empirically associated with the occurrence of LBP (Hagberg et al., 1995) and are not necessarily the direct causes of LBP. These risk factors can directly or indirectly influence the onset and/or course of LBP by affecting the pathological and physiological responses (Hagberg et al., 1995).

Many occupational guidelines and ergonomic checklists have been designed to define and address the issues related to risk factors at occupational sites. The main problem in the utilization of these guidelines is that these risk factors are not independent of one another, and generally, they may act as surrogates for each other (Hagberg et al., 1995). A risk factor can have a direct effect through various mechanisms or it can influence other risk factors by manipulating their relative importance. For instance, the most often cited risk factors of LBP are work intensity (Marras et al., 1995) and poor posture (Punnett et al., 1991). In occupational settings, a task may define a specific physical demand on the body and/or a specific beginning and ending of a movement; i.e. moving an object from point A to point B. However, the choice of the movement pattern is not defined. Depending on the physical demand of the task (e.g., shape of the object, bulkiness), one may choose a specific movement pattern. Depending on the physical demand and load of the object, a safe movement pattern may become unsafe and increase the risk of LBP. Generally, this interaction is not trivial and the workers should be educated and trained to understand these phenomena. Furthermore, the interaction becomes more complex when other factors such as velocity and inertia of dynamic movement are added.

Most of the recent literature pertaining to asymmetrical versus symmetrical loading of the body addresses the issues related to the interaction of load and posture. These studies are important for two reasons: 1) they refer to the two risk factors (i.e., load and posture) that are most frequently cited in the literature as occupational risk factors for LBP; and 2) asymmetrical posture is observed more frequently in industry than symmetrical posture. The aim of this study is to further investigate the interaction of the load and posture. Following is a pertinent description to this study of each of these risk factors and a review of the literature related to asymmetrical lifting.

1.1.1 Load

Mechanical load, as used in the literature, can be divided in three main categories: 1) whether the load is defined externally or internally, 2) whether the magnitude of the load is high or low, 3) whether the duration of the load is short or long. All combinations of these categories can be found in the current biomechanical literature.

The early preventive strategies for LBP were based on reducing the magnitude of external load on the spine. In 1981, the National Institute for Occupational Safety and Health (NIOSH) published a work practice guide for manual lifting (NIOSH, 1981). The guideline was based on the theory that "overexertion injury is the result of job demand that exceeds a worker's capacity" (NIOSH, 1981). The guideline was limited to static and symmetrical load and the duration of the load was not taken into account. The model was revised in 1991 (Water et al., 1993) and the affect of other parameters, i.e. asymmetrical load, was incorporated in the model.

Most recent studies indicate that prevention strategy should be based on the reduction of magnitude of both external and internal loads on the spine. The rationale for this strategy stems from a series of studies that investigated the effect of asymmetrical and symmetrical external load on the coactivation of trunk muscles. These studies (Lavender et al., 1994; Marras et al., 1993) demonstrated that the characteristics of external load (magnitude of load, symmetrical and asymmetrical nature of the load) affect the pattern of coactivation and the magnitude of forces generated by the trunk muscles. The information from these studies has been used to calculate the spinal compression and shear forces due to the net result of internal forces. These models are still in the evolutionary stages (Marras and Sommerich, 1991) and require further improvement. More detailed discussion of these models are presented in Chapters III and IV.

One shortcoming of biomechanical models is that the load exposure duration is not incorporated in the study of load. Knowledge of the magnitude and pattern of external and internal load is important with respect to understanding the risk and mechanism of injury. Biomechanically, the human body is considered as a composite structure and tissues with viscoelastic properties. Failure (fracture or rupture) occurs when excessive energy is being absorbed by the tissue. Energy can be absorbed whenever a load is being applied on the material or the tissue. Manipulation of external load or posture is the net result of internal forces. Many loading schemes are possible in which the duration or magnitude of load is being varied. Sudden exertions (i.e., explosive exertion, falling, cough) are examples of high load with short duration on the soft tissues. Low cocontraction of trunk muscles during exposure to whole body vibration or repetitive tasks in industry is an example of high duration and low load on the tissue. This type of injury mechanism is referred to as a cumulative trauma model of injury where the rates of micro-injury or micro-trauma are faster than the rate of healing.

It has been reported that overexertion causes low back pain in 60% of all low back pain patients (NIOSH, 1981). Overexertion can be due to high load with short duration or low load with long duration. Current statistical information does not provide information with respect to the type of overexertion. However, it is believed that single overexertion (high load with short duration) is probably a less frequent cause of injury and statistical information is more reflective of the cumulative model of injury (Andersson, 1990).

1.1.2 Posture

The importance of a good working posture has long been recognized (Haslegrave, 1994) and has been explicitly or implicitly associated in the literature with work-related musculoskeletal disorders (Hagberg et al., 1995). In spite of a large body of ergonomic literature related to working posture, no clear definition of posture can be found (Haslegrave, 1994). Based on the particular context, i.e., anatomy or biomechanics, posture may be regarded as the configuration of body parts in space or 'quasistatic biomechanical alignment' (Haslegrave, 1994; Rohmert and Mainzer, 1986). In the later definition, posture is used to refer to dynamic movement of the body and the movement pattern of body parts in space.

It is shown that "poor" posture affects a worker's performance and increases fatigue, induces pain, reduces efficiency, and increases the risk of injury. Based on a field study, Corlett and Richardson (1981) discussed the relationship between worker posture and pain, and efficiency of performance. For example, in the case of a welder who has a limited choice of working postures (Corlett and Bishop, 1976), while holding the pieces together, or an operator who has to bend over a work surface to see and manipulate a workpiece. These poor adapted postures could lead to postural stress, pain, and/or discomfort fatigue. In these cases, the pain or discomfort rather than functional capacity may often be the limiting factor to performance.

Epidemiological and biomechanical studies have found that a combination of high external load and "poor" movement patterns cause a high internal load on the spinal structures and increases the risk of low back pain. "Poor" movement patterns consist primarily of bending and/or twisting of the trunk. Bending occurs during reaching and lifting of an object from a low or high surface. Twisting of the trunk is mostly the result of inadequate workplace design. Often these movements are combined and involve carrying an object. Excessive bending and twisting of the trunk have been related to reported low back pain in industry (Magora, 1973). The frequency of the movement, the pattern, and the weight of the object are suggested to be related to low back pain (Ayoub and Mital, 1989; Bigos and Battie, 1991; Chaffin and Andersson, 1984; Magora, 1973).

Punnett et al. (1991) studied the effect of non-neutral trunk posture among automobile assembly workers. They found that back disorders were associated with mild trunk flexion (Odd Ratio 4.9, 95% confidence interval (CI) of 1.4-17.4), severe trunk flexion (Odd Ratio 5.7, 95% CI 1.6-20.4), and trunk twist or lateral bending (Odd Ratio 5.9, 95% CI 1.6-21.4). A higher risk was reported for jobs with daily exposure to two or three non-neutral postures than those with exposure to only one posture.

An important part of ergonomic job evaluation is to utilize biomechanical models to evaluate the safety of job demands on the worker, based on the posture adapted during task performance. The NIOSH lifting guideline (NIOSH, 1981) and the predictive strength model developed by the University of Michigan (Chaffin and Andersson, 1984) are examples of such models for estimating load on the lumbar spine. The safety of a load on the lumbar spine is predicted by anthropometic data and load characteristics (i.e., weight, distance of load from body). There are two main problems with these models: 1) they can only analyze static or quasistatic postures, and 2) they only analyze the posture based on information available from the actual performed task. However, there is no method of predicting possible safe postures for a given load and specific environmental constraints, e.g., workplace layout and available work space.

Ergonomic literature is significantly lacking in evidence of how a working posture is being adapted by an individual in real work situations, and whether these are "good" or "poor" postures (Haslegrave, 1994). The prediction of posture is important in creating a realistic human model that is to be used in forceful tasks environment (Haslegrave, 1992). Experienced operators are likely to choose very different postures from those adopted by trainees or people not normally employed in heavy manual jobs (Haslegrave, 1992). In most ergonomical job evaluations, the posture is recommended based on the workspace, demand of the job, and biomechanical factors. However, individual posture is the end result of many factors other than biomechanical and ergonomic factors that we are not able to predict or comprehend fully, e.g., comfort, perceived risk of injury or fall (Haslegrave, 1992). Haslegrave (1992) investigated underlying factors for adapted posture during force exertion tasks (force exertion during standing, kneeling, and working overhead). The author reported that posture was affected by the work location, reach distance, and the environmental constraints that were imposed by the layout of workplace. Haslegrave (1992) found a small change in posture constraint resulted in a large degree of variability in the adaptation of posture and a considerable change of strength.

1.1.3 Asymmetrical Load and Posture

Asymmetrical load (asymmetrical task) refers to a situation where the external load on the body is not equal regarding the mid-sagittal plane (Lavender et al., 1992). A common example of asymmetrical loading of the body is carrying, pushing, or pulling with one hand, or a task that requires twisting and lateral bending of the trunk while holding an object. Drury et al. (1982) reported three types of asymmetrical movements in his survey of 2,000 box handlers in a variety of industrial jobs. Trunk twisting was the most common type of asymmetrical movement and occurred during 80% of the tasks. The second type of asymmetrical trunk movement occurred during lifting or carrying a box like object with an asymmetrical hand position. A third type of asymmetrical movement was associated with objects that are unevenly loaded. Common examples of this include carrying containers of liquid (Ayoub and Mital, 1989), boxes that are loaded unevenly, i.e. the center of mass of the load lies outside of mid-sagittal plane (Mital and Fard, 1986), or objects that have an uneven shape.

Some researchers suggests that asymmetrical loading of the body is due to inadequate job design and should be eliminated whenever possible (Garg and Badger, 1986). Ayoub, (1978), suggests that asymmetrical trunk movement is often a natural response of the worker to reducing the muscular effort. If so, the occurrence of asymmetrical movements are often independent of job design and is mainly due to inadequate work technique.

A review of the recent literature suggests several reasons why asymmetrical exertion may increase the risk of injury.

1.1.3.1 Change in the Geometric Alignment of Muscles

The efficiency of the trunk musculature in the force production is affected by trunk twisting and or bending. For instance, depending on the degree of trunk axial rotation, the length-tension relationship of the muscles may change and the ability of the muscles to produce forces will be affected (Pope et al., 1987). Depending on the degree of trunk rotation, the angle of moment with respect to the lever arm of the muscle changes (Pope et al., 1987). The combined effect of these two phenomena will affect the efficiency of trunk muscles in generating force. Moreover, the combined effect of these factors are more pronounced during trunk rotation (McGill and Hoodless, 1990) or trunk rotation combined with trunk bending along the other trunk axis. In the case of pure rotation, it has been shown that trunk twisting affects the centroid, locations, and the orientations of lines action of all trunk muscles (Tsuang et al., 1993). They found that most changes in muscle geometry occur in the first 25 degrees of trunk twisting, and its effect is greater for rather than lower lumbar vertebrae. In some cases, the changes in moment arms were as great as twofold.

The effect of trunk twisting on muscle performance has been demonstrated by the quantification of maximum strength (McGill and Hoodless, 1990). Garg and Badger (1986) reported a significant decrease in static strength as the angle of asymmetrical lifting increased. Mital and Fard (1986) have shown an 8.5% decrease in lifting capability after a 20 cm shift in the central gravity of a box. It was further observed that lifting an asymmetrical load with a 20 cm offset was physically as demanding as lifting a symmetrical, but considerably larger box.

1.1.3.2 Increased Muscle Coactivation

Coactivation of musculature may occur to increase the stability of a joint at higher exertion events. The coactivation of trunk muscles increases as the load increases (Lavender et al., 1994). In the case of asymmetrical movements of the trunk, Marras and Mirka (1990) found that almost all the muscles were affected by acceleration, velocity, and asymmetry of movement. They reported significant coactivation among trunk muscles, and coactivation was different depending on whether trunk velocity or trunk acceleration was manipulated. They observed that a minimal increase of angular torque in the transverse plane was associated with an increase of up to 50% of maximum electromyography (EMG). Lavender et al. (1992) reported an increase in the coactivation of eight trunk muscles in response to asymmetrical loading of the trunk. The pattern of coactivation was different when the directions of the applied load were anterior and exceeded more than 45 degrees from the mid-sagittal plane.

1.1.3.3 Increased Load on the Motion Segment

The basic mechanical unit of the spinal column is the motion segment or spinal functional unit. A motion segment consists of two adjacent vertebra and their intervening disc and ligamentous tissues (Schultz and Miller, 1991). The response of a motion segment to different loading situations has been studied theoretically and experimentally (Panjabi et al., 1989; Shirazi-Adl, 1994; Shirazi-Adl, 1989; Shirazi-Adl and Drouin 1987; Shirazi-Adl et al., 1984). Torsion is mainly resisted by the articular facets and disc annulus fibrosis while the ligament contribution is insignificant (Shirazi-Adl, 1989). A finite element model of contact loads on facets of an L2-L3 lumbar segment indicates that lifting, twisting, and lateral bending of the trunk significantly increase the forces transmitted through the compression facet (Shirazi-Adl et al., 1984). Moreover, twisting and lateral bending were shown to increase the vulnerability of the lumbar disc to rupture (Shirazi-Adl, 1994; Shirazi-Adl and Ahmad, 1991, Shirazi-Adl and Drouin, 1987). The finite element model of heavy lifting shows that asymmetrical lifting increases the risk of structural failure of both the disc annulus and the facet joint as compared to symmetrical lifting (Shirazi-Adl, 1994; Shirazi-Adl, 1984). This result provides some explanation for the epidemiological studies that indicate a higher risk of back injury among jobs that involve more frequent asymmetrical loading of the trunk (Andersson, 1981; Marras et al., 1993; Punnett et al., 1991; Riihimaki et al., 1989).

1.1.3.4 Increased Compression Load on the Disc

The intervertebral disc is a major load-bearing element of the spine in lateral and anterior shear and axial compression forces. Biomechanical models have traditionally been utilized to calculate the compression and shear forces on the disc (Nachemson and Morris, 1964). Theoretically, compression should increase with increased coactivation of the muscles. Higher muscle coactivations can occur either as a result of a change in the angle of net external moment or with an increase in the magnitude of external load.

In vitro, Farfan et al. (1970) showed that the trunk torque and angles required for disc failure are easily exceeded in daily tasks involving twisting. For a given flexion angle of the trunk, Andersson (1985) reported that trunk rotation along any axis of the trunk increases the disc compression. Other researchers found a direct effect of asymmetrical muscle exertion on lumbar discal compression (Granata and Marras, 1993; Mirka 1988). McGill and Norman (1993) reported that support of 50 Nm load in extension imposes about 800 N of compression on the lumbar. The same load imposes compression of 2500N in twisting and 1400N in lateral bending of the trunk. The variation in the compression load was attributed to a difference in the coactivation of trunk muscles, and changes in the moment arms (McGill and Norman, 1993).

1.1.3.5 Increased Abdominal Pressure

Intra-abdominal pressure (IAP) is believed to play an important role in spine biomechanics and lifting. There are controversial reports concerning the quantitative value of intra-abdominal pressure and its role (Gracovetsky et al. 1985; Marras et al. 1984; Schultz et al. 1982). Further discussion related to the role of IAP can been seen elsewhere (Cresswell et al., 1994; Cresswell and Thorstensson, 1994; McGill and Norman, 1993; Pope et al., 1991). With respect to asymmetrical lifting, Marras and Mirka (1990) reported an increase in abdominal pressure during asymmetrical trunk angular acceleration. However, the increase was not statistically significant.

Many investigators attempted to determine and compare interactions between the abdominal musculature and intra-abdominal pressure during different trunk loading conditions (Cresswell and Thorstensson, 1994; Gracovetsky et al., 1985; Marras et al. 1984; Schultz et al., 1982). These studies mainly investigated the role of IAP during specific tasks and suggested an association between IAP and the direction and magnitude of the load (Cresswell et al., 1992; Cresswell and Thorstensson, 1994), deceleration and acceleration (Cresswell and Thorstensson, 1994). However, no unifying hypothesis exists to explain the role of IAP for a wide variety of movement tasks (McGill and Sharratt, 1990).

1.1.3.6 Psychophysical Effects

Several authors (Drury et al., 1989; Garg and Badger, 1986; McGill, 1991) studied the effect of asymmetrical work on psychophysical factors. The primary comparison concerns the maximum acceptable weight during symmetrical vs. asymmetrical lifting tasks. Garg and Badger (1986) reported a significant decrease in the maximum acceptable weight as the angle of asymmetry increased during a lifting task. They suggested correction factors of 12, 21, and 31% for the maximum acceptable weight during 30, 60, and 90% of asymmetric lifting respectively.

Ferguson et al. (1992) studied the effect of dynamic asymmetrical tasks on trunk motion characteristics. Compared to maximum static strength, a 21% reduction of strength was observed during dynamic asymmetrical lifts.

1.1.3.7 Physiological Effects

Asymmetrical manual material handling increases the energy cost and heart rate (Garg and Badger, 1986). However, it is not clear if the increased physiological factors are statistically significant. Mital and Fard (1986) found no significant effect of asymmetrical lifting on the heart rate and oxygen consumption during lifting of the maximum acceptable weight. They concluded that the reduction of the maximum acceptable weight was such that the levels of physiological response remained unchanged.

In summary, the review of the literature indicates that a variety of risk factors have been associated with low back pain. These risk factors interact with each other and their interaction possibly defines the risk LBP. The interaction of these variables needs to be studied in order to better understand of LBP in industry. Asymmetrical activities, i.e., asymmetrical exertion and asymmetrical posture, are the most common type of activities in industry. These activities have been associated with the risk of low back injury. It is important to further investigate the effect of asymmetrical activities and the consequences of these activities on spinal forces and the risk of low back injury.

1.2 Aim of the Study

The literature review suggests that the epidemiological data shows an association between exposure of twisting and lateral bending of the trunk, and the reported incidence of low back pain. However, the exact nature of this relationship is not known. Biomechanical models are used to investigate the relationship of these risk factors to mechanical load on the spine and occurrence of low back pain. The main purpose of these models is to study the effect of external load on the internal structures and to estimate the magnitude of the load on these structures.

Most of our knowledge regarding internal forces during asymmetrical exertion is based on studies that investigated internal spinal load during dynamic activities. The internal loads were estimated via EMG-driven models of the trunk. High spinal forces during dynamic asymmetrical activity shown by these studies is partially due to change of posture and thereby changes in muscle alignments, and partially due to the changes in net resultant moments. EMG-driven models are sensitive to the accuracy of the anatomical description of muscles, i.e., muscle cross-sectional area, lever arm, and line of action. No method is available to correct or predict actual changes in the lever arm and the centroid of muscle during dynamic trunk activity. Previous EMG-driven models of the trunk assumed constant anatomical values for the trunk muscles during dynamic activity. The validity of this assumption is not known.

The aim of this study is to quantify the activities of ten trunk muscles by using surface and wire electromyography during maximal and submaximal isometric exertion under symmetrical and asymmetrical trunk exertion. Symmetrical exertion is defined as pure planar flexion and extension of the trunk in the sagittal plane. Asymmetrical exertion was simulated by combined exertion of the trunk along the sagittal and transverse planes. These planes are selected based on their prevalence in industrial tasks and low back injuries.

Several measures have been taken to reduce the error in force prediction. The study is limited to isometric exertion in order to eliminate the error due to changes in the muscle-length-tension curve and the geometry of the muscle. A repeated measure design is used in this study to eliminate the within-subject error. Submaximal exertion was selected based on percent of Maximal Voluntary Exertion (MVE), and muscle parameters in the mathematical model are measured from a CT-scan of the subject's trunk.

1.3 Hypothesis

The primary hypotheses of this study are:

1. The mean of RMS-EMG of the ten selected trunk muscles will change significantly with the orientation and magnitude of the net resultant moments of the trunk.
2. The result of the EMG-driven models will be significantly affected by assumptions made with respect to the EMG-force relationship of the trunk muscles.
3. The result of the EMG-driven models will be significantly affected by the assumptions made with respect to the geometrical description of the trunk muscles.

The rationale for the first hypothesis is based on the anatomical and biomechanical observation of the trunk during asymmetrical trunk exertion. The coactivation of the trunk muscles produces a net trunk moment along all three axes. In the case of planar flexion, the rectus abdominus muscle is considered the primary mover. For extension, the erector spinae muscle is considered the primary mover. In both cases, the muscles are aligned within the plane of exertion. In axial rotation of the trunk, no single muscle is defined as the primary mover. The rotational trunk exertion is the net resultant of the coactivation of several muscles. Combined exertion requires some muscles to fulfill more than one function, due to the lack of primary trunk muscles for rotation and the limitation of the number of trunk muscles. This provides the rationale for the problem of muscle allocation during an asymmetrical task.

The rationale for the second hypothesis is based on the physiological and biomechanical properties of the trunk muscles. Both linear (Lippold, 1952; Woods and

Bigland-Ritchie, 1983) and non-linear (Komi and Buskirk, 1970, Zuniga and Simons, 1969; Lawrence and De Luca, 1983) EMG-force relationship has been presented in the literature. EMG-driven models have been used to estimate spinal forces. These models assume a linear EMG-force relationship, and the relationship is explained by the linear line that has zero intercept. Non-zero intercept can be assumed in these models. No study investigated the sensitivity of EMG-driven models to different assumptions of the EMG-force relationship.

The rationale for the third hypothesis is based on the anatomical and functional role of the internal oblique muscle. In EMG-driven models, depending on the role of the muscle as defined by the investigator, the internal oblique muscle has been defined to produce a flexion or extension moment in sagittal plane. No study investigated the effect of the different formulation of internal oblique muscles on results of EMG-driven model.

1.4 Research Question

Two sites of have been suggested in the literature for monitoring the myoelectrical activity of the internal oblique muscle (IOB) of trunk, the anterior triangle (Basmajian and De Luca, 1985; Floyd and Silver, 1950; Snijders et al., 1995), and the posterior triangle (O'Rahilly, 1986; Williams et al., 1989). These triangles represent the area where the internal oblique is not covered by the external oblique muscle, and surface electrodes can be used to investigate the function of this muscle. Placement of electrodes in previous studies (Marras and Mirka, 1990; Marras and Mirka, 1992) at the posterior triangle area suggests that the IOB muscle acts as trunk extensor. This is contrary to the previous studies (Basmajian and De Luca, 1985; Floyd and Silver, 1950; Snijders et al., 1995) that monitored myoelectrical activity of the IOB muscle at the anterior triangle and found that this muscle acts as trunk flexor. To answer the research question, this study will investigate the functional behavior of the internal oblique muscles during pure and combined trunk exertion, using wire electrodes.

1.5 Outline of the Study

This thesis is presented as the form of four manuscripts. Chapter II investigates the EMG activity of 10 trunk muscles during pure and combined trunk exertion. The aim of this section is to test the first hypothesis of this study. Chapter III the describes formulation of an EMG-driven model of the trunk for prediction of spinal force. Chapter IV investigates the effect of anatomical geometry of trunk muscles on the results of an EMG-driven model. The objective of Chapter IV is to test the third hypothesis. Chapter V contains a summary of the studies and recommendations for future research. Chapters II-IV includes similar description of testing procedures. The information is repeated in order to keep the chapters individually independent.

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Chapter II

Combined Sagittal and Transverse Isometric Exertion During Upright Standing Trunk Performance

2.1 Review of the Literature

The understanding of the etiology and methods for prevention of low back pain have been a global concern shared by researchers and professionals of many disciplines for the last few decades (Ayoub and Mital, 1989). The main reason for this concern is the extensive human suffering and economical cost caused by low back pain (LBP).

Despite numerous epidemiological and biomechanical studies, the exact etiology of low back pain remains unclear. It seems, however, that there is a general consensus on possible injury mechanisms leading to low back pain. This is based on a statistical or experimental association between specific risk factors and the occurrence of low back pain (Hagberg et al., 1995) and are not necessarily the direct causes of LBP.

Epidemiological and biomechanical research has shown the relationship between activities that cause large mechanical loads on the low back, risk of injury, and low back pain (Andersson 1985; Keyserling et al.; 1988, Van Dieen, 1993). Controversy exits as to which tissue is involved and which property of the tissue is the cause of failure. Practically every structure has been at one time postulated to be the locus of the LBP (Van Dieen, 1993). Based on a biomechanical understanding of the movement and complexity of the interaction between involved structures, it is very likely that damage to any structure will lead to an alteration of the movement pattern and overloading of secondary structures (Panjabi et al., 1989).

One of the main challenges for biomechanical studies of LBP is to investigate the mechanisms and conditions where mechanical load can become a risk factor and estimate the associated risk. Generally, the testing conditions and variables used in these investigations are based on epidemiological data. A review of the literature (see Chapter I) suggests that a combination of heavy external load and asymmetrical loading of trunk (i.e., bending and/or twisting) may cause a high internal load on the spinal structure and may increase the risk of low back injury and pain.

Since no direct noninvasive method of measurement of internal load of each structure is available for in vivo study, indirect measurement techniques are often used. Among indirect measurement techniques, electromyography (EMG) provides comprehensive information with respect to internal load. EMG is thought to have a proportional relation to the force produced by the muscle. However, the nature of the EMG to force relationship is not clearly known.

With respect to testing conditions used to study asymmetrical load conditions, EMG studies of trunk muscles can be divided in two main categories: static and dynamic. Typical dynamic testing conditions consist of the investigation of the EMG activity of trunk muscles during dynamic movement of the trunk along a single trunk axis, where the resistance, the velocity, or the acceleration were being manipulated. Marras and Mirka (1990) describe the behavior of 10 trunk muscles in response to changes in trunk velocity and trunk acceleration. They reported a significant coactivation among trunk muscles, and the coactivation was different depending on whether trunk velocity or acceleration was manipulated. McGill (1992) demonstrated that change in lumbar lordotic posture (i.e., hyperlordotic or hypolordoctic posture) reduces the dynamic trunk twisting torque. Independent of lordotic posture, dynamic torque was shown to be a function of the direction of torque exertion.

Dynamic studies represent a more realistic testing condition than static testing. Dynamic studies allow both realistic human performance and mechanical loading, with the shortcoming that the response of the central nervous system (CNS) to a single variable cannot be isolated. In these studies, the response of the CNS is always confounded with postural effects; changes in the geometry of the muscle and the length of the soft tissues. Twisting and bending changes the lever arm and the length of the muscle. Tsuang et al. (1993) showed that the centroid of the muscle changes during trunk twisting movement. In dynamic biomechanical models, only the EMG length modulation factor of the trunk muscles can be accounted. In reality, the EMG to force relationship also exhibits dependency on the muscle length and the rate of muscle contraction. In contrast, static testing condition, although representing less realistic conditions, can offer a better opportunity to study the response of the CNS to isolated variables.

With respect to asymmetrical trunk exertion, the studies concentrated on two main questions; the effect of asymmetrical exertion within asymmetrical posture (Lavender et al., 1993; Marras and Granta, 1995; Pope et al., 1987) and the effect of asymmetrical exertion within symmetrical posture (Ladin et al., 1989, Lavender et al., 1992). Pope et al. (1987) demonstrated that pre-rotation of the trunk during twisting effort will affect the torque production of the trunk. The axial rotation torque is less in the direction of twisted posture and more in the opposite direction. Similar results were reported by others (Lavender et al., 1993; Marras and Granta, 1995).

Marras and Granta (1995) studied trunk twisting during both isometric and dynamic exertion. They observed that twisting exertion is highly coupled with sagittal and lateral moment. The coupled moments generated were about 20% of MVE in the sagittal plane and 75% of MVE in the lateral plane. Furthermore, they observed less axial rotation moment during dynamic exertion compared to isometric rotation moments.

Lavender et al. (1993) quantified the activation of eight trunk muscles as subjects maintained a twisting posture and resisted external bending moments. The external moment of 20 and 40 Nm was directed in a transverse plane and the direction of moment was from the anterior to the posterior direction with an increment of 30 degrees. The trunk muscles were significantly affected by the interaction of the moment direction and moment magnitude. Furthermore, Lavender reported that the region of the loading plane in which the trunk muscles are active, the "switching curve" as it was previously defined by Ladin et al. (1989), will likely shift during twisting posture of the trunk.

Lavender et al. (1992) investigated the affect of asymmetrical exertion on symmetrical upright posture. They quantified the activation of trunk muscles while the torso was loaded with external moments of 10 to 50 Nm from seven directions. They reported large changes in muscle recruitment due to asymmetry, but only small changes in muscle activity due to increases in the magnitude of moments.

No study investigated the effect of combined transverse and sagittal exertion. In an attempt to better understand the pattern of muscle recruitment under asymmetrical loading, the aim of this study is to investigate trunk performance during combined exertion without the artifact of posture and changes in geometry of trunk muscle.

2.2 Hypothesis

The primary hypothesis of this study is that the mean of the normalized root mean square of electromyographical signal (NRMS-EMG) of the ten selected trunk muscles will change significantly with the orientation of the net resultant moments (pure and combined sagittal and transverse moments) of the trunk.

The rationale for the hypothesis is based on the anatomical and biomechanical observation of the trunk during asymmetrical trunk exertion. Based on the orientation and line of action of the muscle, the force generated by a muscle can contribute to the net moment of the trunk along the trunk axis. In response to physical demand, the CNS utilizes these resources (i.e., the trunk muscles) to exert force along one or more axis. However, the strategy for these recruitments is not well understood.

It is generally believed that muscles will be recruited fully during isometric contraction. Due to the lack of primary muscles for twisting of the trunk and the limitation of the number of trunk muscles, one expects that a combined exertion requires some muscles to fulfill more than one function. This provides the rationale for the problem of muscle allocation during the performance of an asymmetrical task.

Furthermore, if the trunk is thought of as a purely linear mechanical system, one would expect that maximum trunk exertion during combined exertion would be equal to the vectorial sum of maximum voluntary exertion during uniplanar trunk exertion. The validity of this assumption has been tested in our study by comparing the torque produced during combined exertion with a vectorial sum of uniplanar exertion.

2.3 Method

Eighteen subjects between the ages of 20-40 years with no history of low back pain for the last six months were recruited for the study. The subjects were recruited based on advertising in the New York University campus and the Hospital for Joint Diseases, New York city. The study was approved by the Internal Review Board of the Hospital for Joint Diseases (IRB No. 22-103) and New York University.

2.3.1 Instrumentation

A multichannel data acquisition system was developed in-house to simultaneously collect EMG signals and mechanical data from the three dimensional dynamometer. The system is currently able to collect data from 16 channels of EMG and 16 channels of mechanical data through an analog to digital (A/D) board (ATMIO-64F National Instrument, TX) at a sampling rate of 1000 Hz. The system consists of three parts: a B200

Isostation, an EMG system, and a visual feedback system.

2.3.1.1 B-200 Isostation

The B-200 Isostation (Isotechnologies, Inc., Hillsborough, NC) was used to measure the torque generated during maximal and submaximal trunk exertions. The device is a triaxial trunk dynamometer that simultaneously measures angular torque and about three exertion axes. The system was interfaced with an IBM-PC 486 computer. The validity and reliability of the device, and its efficacy for monitoring isometric trunk torques, has been previously reported in the literature (Ross et al., 1993; Schmitz, 1992; Tan et al., 1993).

2.3.1.2 EMG System

The EMG signals were collected by a 16 channel EMG-system (Innovative Computer Solutions, Fresh Meadows, NY). The surface electrodes were connected to the EMG-system for amplification and filtering (high pass filter at 25 Hz and low pass filter at 2000 Hz). The EMG-system was interfaced with A/D board with 1000 Hz sampling rate.

2.3.1.3 Feedback System

The Labview software (Version 3.0, National Instrument, Austin, TX) was used to collect data and provide real time visual feedback regarding the torque measured by the B-200 Isostation. During each trial, the program displayed a feedback screen which consists of a real time display of the subject's generated sagittal and transverse torque and ± 5% of the expected exertion target (see Figure 2.1). The desired level of exertion for each axis was specified before each trial.

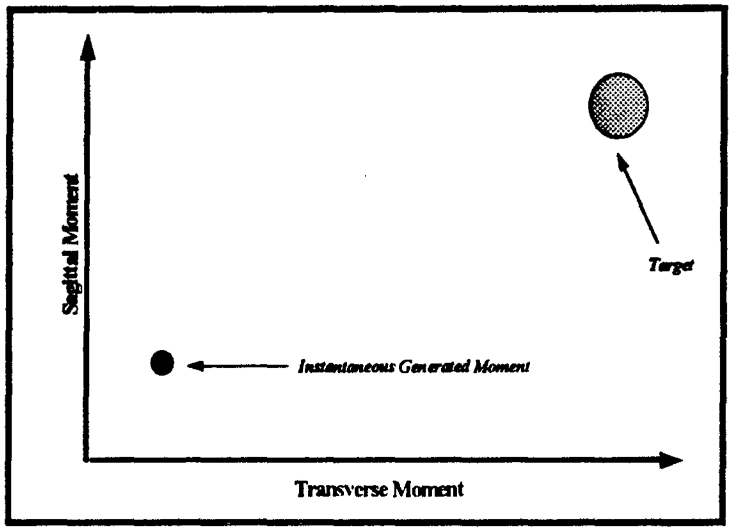

Figure 2.1. The visual feedback screen provided for the subject during each trail. The horizontal axis exhibits right and left rotation transverse moment. The vertical axis exhibits flexion and extension moment in the sagittal plane. The small black circle represents the real time display of torque being exerted. The large gray circle indicates the target area and shows ± 5% of target torque.

2.3.2 Procedure

All subjects were interviewed and screened for the study. The screening was done by means of a questionnaire. Each subject signed the consent form (Appendix E) and filled out a habitual physical activity form (Baecke et al. 1982, Appendix B) and a demographic and medical questionnaire (Appendix A). The demographic and medical questionnaires were reviewed to ensure that the subject satisfied the inclusion criteria and had no history of low back pain in the last six months, nor any neurological and physiological conditions that would make him/her unsuitable for this study.

2.3.2.1 Step 1. Wire Electrode Placement

To insure the safety and accuracy of wire electrode placement, wire electrodes were placed using a needle biopsy procedure under CT-scan imageing. Upon the completion of the questionnaires, the subject was scheduled for a CT-scan. The trunk scan was taken using a Computed Tomographic unit at the Hospital for Joint Diseases Orthopaedic Institute (CT 9800 HiLight). Before and during the testing, the subject was instructed to relax but not to move or change posture. The subject was in a supine position. The L3/L4 level was accurately found from the sagittal plane image. At this level, markers were placed on the skin. A CT-Scan was then taken at the L3/L4 level in the transverse plane. Based on marker placements and the location of the right and left internal oblique (IOB) muscle, the site of insertion of the wire electrode was selected by a board certified radiologist (see Chapter IV for more details). The insertion of wire electrode was done under CT-scan into the middle third region of the right and left IOB muscle. More detail information for site of wire insertion is presented in Chapter IV.

2.3.2.2 Step 2. Surface Electrode Placement

Surface electrodes were used to detect the myoelectric activity of four bilateral trunk muscles. The electrodes were bipolar, pre-gelled surface electrodes (Classic Medical, Muskego, WI) with an inter-electrode separation of 20 mm. The electrode placement sites were based on anatomical considerations, palpation of muscles during contraction, and a review of previous studies (Schmitz, 1992; Tan et al., 1993). The following bilateral trunk muscles and sites were used in this study.

Two pairs of electrodes were placed bilaterally on the lumbar spine at the L3 level to record activity from the medial erector spinae (ES) muscles, as shown in Figure 2.2 (Gabriel, 1992; Pope et al., 1987; Ross et al., 1993). The belly of the muscle was verified by palpation of the muscles while the subject was manually resisting trunk in extension in slightly forward flexed posture. Two pairs of electrodes were used bilaterally on the lower thoracic spine at the T12-L1 region to monitor the activity of the latissimus dorsi (LAT) muscles. The exact location of the electrode placement on the muscle belly was determined by palpation during manually resisted extension of the upper arm and shoulder in a neutral standing posture with flexed elbow ninety degrees . The T12-L1 level was selected due to practical considerations during the testing and the possible interference of the thoracic pad of the B-200 Isostation.

Four pairs of electrodes were used to record activities from the abdominal muscles (Figure 2.3). Two pairs of electrodes were placed bilaterally midway between the iliac crest and the lower border of the rib cage, over the external oblique (EOB) muscles. The choice of position was based on anatomy and reported sites in the literature (Mayhew et al. 1983, Schultz et al., 1987). The electrodes were oriented in the direction of the external oblique muscle fibers (Schultz et al., 1987).

Two pairs of electrodes were placed bilaterally over the rectus abdominus (RA) muscle. They were placed at the same level as the electrodes for the oblique abdominal muscles and about 3 cm lateral to midline (Schultz et al., 1987).

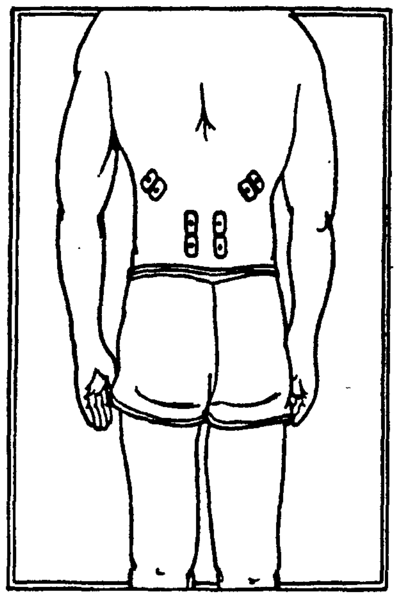

Figure 2.2. Location of electrodes on posterior side of the trunk

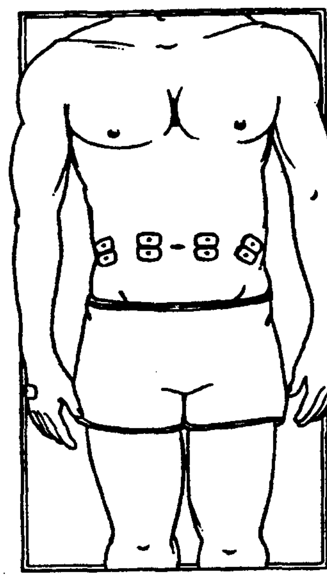

Figure 2.3. Location of electrodes on Anterior side of the trunk

2.3.2.3 Step 3. Maximum Voluntary Exertion (MVE)

The participant was strapped to the B-200 Isostation. The positioning was upright standing and the method of constraining the chest, pelvis, and thigh are explained in detail in the B-200 User's Manual (Isotechnologies, 1988) and previous studies (Gabriel 1992; Ross et al. 1993; Schmitz, 1992; Tan et al., 1993). The subject performed two trials of trunk MVE randomly in four directions: flexion, extension, and right and left axial rotation. The sequence of these test was randomized by the computer. During these MVEs, the secondary axes were unlocked. The subject was asked to perform at a maximum effort for three seconds with no jerky exertion. Two minutes of rest separated each test to avoid fatigue. If the MVE of the two trials varied more than 10% , the subject was tested for a third MVE and the two trials with consistent values were selected. Based on the average of the two trials of MVE, the values for 30, 60, and 100% of MVE for each direction of the sagittal and transverse planes were calculated individually.

2.3.2.4 Step 4. Randomization

For each test, the computer randomly selected a test from all possible combinations of 0, 30, 60, and 100% of MVE of flexion, extension, and rotation. Figure 2.4 shows the selected testing conditions. Figure 2.5 present the angle notation used in this study for angle of net resultant moments.

No evidence exist to indicate variation in patterns of muscle recruitment between left and right rotation. To reduce the number of testing conditions, the risk of fatigue, and patient discomfort, only exertion to the right side is considered in this study. No load condition, zero condition along each axis, was repeated more than three times. However, these tests were part of the randomization process. The zero condition served as a means to calculate the baseline EMG for the normalization of the data.

2.3.2.5 Step 5. Training

The subjects were trained for the testing protocol. The following information was explained to the subject:

1) During the testing, the subject should look at the monitor (VDT) to evaluate his/her performance. The black ball on the screen will show the real time feedback of torque with respect to the selected planes, (Figure 2.1). The goal is

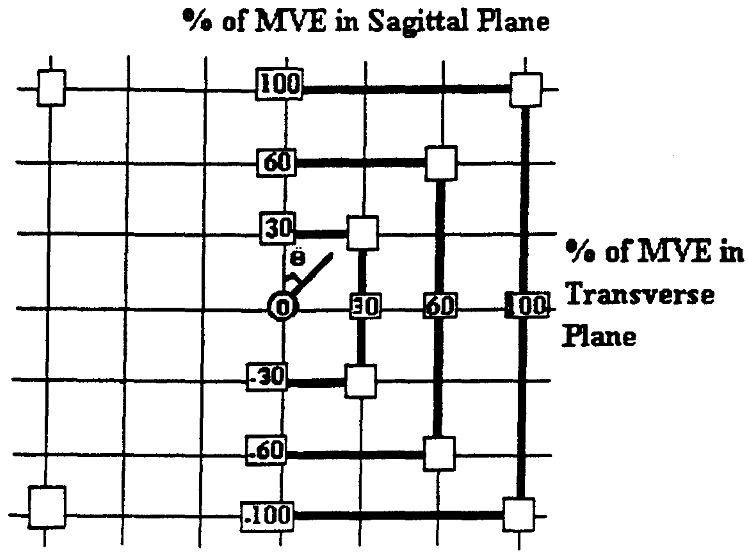

Figure 2.4. Testing conditions used in this experiment. The boxes are showing the testing conditions. The value of angle (θ) express the net resultant moment of the trunk (see figure 2.5 for values of this angle).

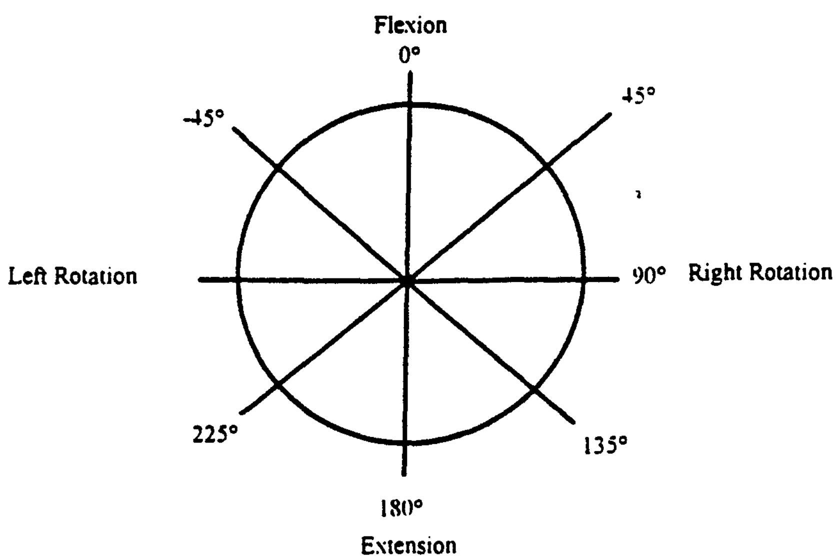

Figure 2.5. The angle notation (θ) used in this study.

to generate moments in either of the two planes that move the ball to the target. The position of the target is based on the selected exertion level and the task demand.

2) The subject should try to reach the target and stay within the target during the testing period for at least 3 seconds continuously.

3) The subject should try to reach the target as quickly as possible, since the maximum data collection time is 10 seconds. However, it is not desirable to produce a jerky exertion. The purpose of this test is to reach the target as smoothly as possible.

4) The subject was informed that should the task demand be exceeded, the following procedure would be implemented. If the subject has exceeded the reference torque value, the ball will stop at the extreme corner of the over-exerted axis. During all the testing conditions, the lateral plane is unlocked. However, to maintain the upright posture, the movement in the lateral plane will be monitored. If the subject exceeds ±5 degrees of lateral bending, the computer will generate audio feedback and the subject will need to correct his/her upright posture.

5) Training is done for several testing conditions. Whenever the subject feels comfortable with the setup and testing procedure, he/she can notify the examiner to start the actual testing protocol.

2.3.2.6 Step 6. Testing

Before each test, the feedback screen was displayed on the terminal and the subject was asked to practice reaching the target area. Based on this performance, verbal feed back was provided. The practice was continued until the subject was comfortable with the task and indicated that he/she was ready to start the test.

Based on the randomization procedure described in step 4, the examiner set the exertion level of each axis and the subject was asked to perform the randomized test sequence. The examiner set the levels of task demand without notifying the subject. Between each testing condition, the subject rested for 2 minutes.

2.3.3 Data Analysis

The following steps were taken in data reduction and data processing for statistical analysis.

2.3.3.1 Step 1. Data Reduction

Each data file contained 10 seconds of data for three axial torques and 10 raw EMG channels. Based on Equation (1), a control index (IC) was calculated for each data file. A three second moving average was used to find 3 seconds of the data where the subject has lowest index value. The low value of this index represents the high controllability and capability of performing close to target torque.

$$\text{Control Index} = \frac{\sqrt{\left(Sag_{Torque} - Sag_{Targ}\right)^2 + \left(Tran_{Torque} - Tran_{Targ}\right)^2}}{\sqrt{\left(Sag_{Targ}\right)^2 + \left(Tran_{Targ}\right)^2}} \tag{1}$$

Where :

Sag_{Torque} and $Tran_{Torque}$= The performed sagittal (Sag) and transverse (Tran) torque
Sag_{Tag} and $Tran_{Tag}$ = The sagittal (Sag) and transverse (Tran) target torque

The data within the selected 3 second window was used for all further data analysis and measurement of torque and EMG within each trial.

2.3.3.2 Step 2. Inception of EMG Signal

To insure the quality of the EMG signal, the raw EMG signal derived from the 3 second selected window was displayed on the computer screen. The mean and standard deviation of absolute value of the selected window was calculated. The mean ±2.5 standard deviation was used as the criterion for selecting an abnormal spike, i.e., noise or movement artifact. If the spike exceeded this limit, the spike was replaced by the mean of that data set. In cases where a continuous series of data points exceeded the limit, the complete section of data was eliminated from the data set and the process was repeated to find other abnormal spikes. In this case, this procedure resulted in a window of less than 3 seconds data for further analysis.

2.3.3.3 Step 3. Normalization of Data

To normalize the EMG data, the mean of Root Mean Square(RMS) of the EMG signal was calculated for a selected 3 second window. Based on previous recommendations (Redfern, 1992), the time constant of 50ms was used for RMS calculation.

The normalization procedure in this study was based on studies by Lavender et al., (1992); and Seroussi and Pope, (1987), and is as follows:

$$NEMG_i = \frac{Task\ EMG_i - Rest\ EMG_i}{Maximum\ EMG_i - Rest\ EMG_i} \tag{2}$$

where $NEMG_i$ is the normalized EMG of i^{th} muscle.

Previous studies (Lavender et al., 1992) used the maximum EMG value that was observed during MVE. However, in this study, the highest mean NRMS-EMG values among all the trials was used for the normalization of each muscle.

2.3.3.4 Step 3. Statistical Analysis and Hypothesis Testing

Descriptive statistics (mean and standard deviation) were computed for demographic data, maximum isometric torque, and EMG activity of the ten trunk muscles. Intra-subject variability is reported to be smaller than inter-subject variability (Thorstensson et al., 1985). Therefore, a repeated measures design was used to reduce the effect of between-subjects variability. Furthermore, the repeated measures design requires fewer subjects to achieve a given statistical power (Zar, 1984). In this study, level of statistical significant was chosen at $p < 0.05$.

For the hypotheses testing, the statistical procedure multivariate analysis MANOVA with a repeated measures design (SAS User's Guide, 1985) was used to determine the effect of angle (θ) and the level of exertion on ten trunk muscles, Figure 2.4. If the overall MANOVA was significant at $p<0.05$, a univariate analysis of variance (ANOVA) was performed. The Tukey test was performed on the levels of angle and level of trunk exertion, if the effect of the univariate test was statistically significant.

Two way repeated ANOVA design was used to determine the effect level of exertion on produced torque during combined exertion. The dependent variable was the difference between actual torque produced and target torque as it was defined in Equation (1). The dependent variables are the angle of combined exertion (45 and 135 degree) and three levels of exertion (30, 60, 100% of MVE).

2.4 Results

2.4.1 Participant Characteristics

Table 2.1. Demographic data of participants (n=15).

	Male n=10	Female n=5	Total n=15
Age (years)	28.5 (4)	27.2 (8)	29.8 (6.4)
Weight (Kg)	73.1 (11)	60.0 (4)	69.9 (12.9)
Height (cm)	171.8 (6)	159.6 (7)	168.2 (8.7)

Thirteen males and five females with no history of back pain for the past 6 months volunteered for this study. Data from three male subjects were not included in the study due to incompletion of test, (i.e., two subjects were indisposed and could not continue the test and, one subject accidently pulled off the wire electrodes).

Table 2.1 shows the mean and standard deviation for the age, height, and weight of

Table 2.2. Mean and standard deviation of index of work, sport and leisure for all participants (n=15).

	Male n=10	Female n=5	Total n=15
Work	2.33 (0.57)	2.35 (0.41)	2.33 (0.50)
Sport	3.41 (1.68)	3.66 (0.88)	3.49 (1.43)
Leisure	3.17 (0.44)	3.70 (0.54)	3.35 (0.52)

the male and female participants who were included in the study. The mean age was 28.5 (4) years for males, 27.2 (8) years for females, and 29.8 (6.4) years for all subjects.

An index of work, sport, and leisure time was computed from the habitual physical activity questionnaires for all subjects. Appendix C demonstrates the method of calculating these indices from Baecke et al. (1982). Table 2.2 shows the mean and standard deviation indices for males and females. The scores for index of work, sport, and leisure were 2.33 (±0.50), 3.49 (± 1.43), and 3.35 (±0.52) respectively.

2.4.2 EMG Activity of Trunk Muscles

A total of 15 randomize trunk extersions were perforemed. The total testing time was approximatly 2 hours per subject. The subject was asked to report discomfort if it should occure due to surface and wire electrodes. No subject reported discomfort with electrodes during testing.

The details of the descriptive statistics (mean and standard deviation) for normalized mean NRMS-EMG of ten trunk muscles are presented in Appendix D (Table D.1-D5).

The result of the repeated MANOVA for the main effects of angle and level of exertion and their interaction effect on the mean of NRMS-EMG of the ten trunk muscles are shown in Table 2.3. Based on the Wilks' Lambda, the combined dependent variables were significantly affected by both the level of exertion, $F(20,222)=25.835$, $p<0.0001$, and the angle of exertion, $F(40,422)=28.393$, $p<0.0001$, and their interaction effect,

Table 2.3. Multivariate analysis of variance (MANOVA) for the main effect of exertion and posture on mean NRMS-EMG of ten trunk muscles during upright isometric trunk exertion.

	F	P
Exertion	25.8354	0.0001
Angle	28.3936	0.0001
Exertion * Angle	2.65	0.0001

$F(80,712)=265$, $p<0.0001$.

The repeated ANOVA was performed to investigate the effect of three levels of exertion (30, 60, and 100% of MVE) and five angles of exertion (0°, 45°, 90°, 135° , 180°) on ten trunk muscles. Table 2.4 shows the summary of the results of the ANOVA procedure. The details of the ANOVA statistics and outcome values are shown in Table 2.4.

The results of ANOVA demonstrate that the trunk muscles are significantly affected by the magnitude of exertion ($p < 0.001$), angle of exertion ($p < 0.001$), and their interaction ($p<0.001$). Figures 2.6-2.10 graphically shows the effect by plotting the mean NRMS-EMG of the muscle for each level of exertion. To simplify the angle notation, Figure 2.5 shows the angles with respect to the sagittal and transverse planes that are used in this experiment.

All muscles showed a higher level of activity during 100% MVE followed by 60 and 30% MVE. The result of the post-hoc Tukey test showed that all three levels of exertion are significantly different from each other. Table 2.5 contains a summary of the main effect of the exertion for all trunk muscles.

Figures 2-6 to 2-8 show that all the abdominal muscles, right RA, IOB, and EOB demonstrated higher activity during pure and combined flexion (-45°, 0°, 45°), compared to pure and combined extension (135°, 180°, 225°).

The left EOB and right IOB had a similar pattern of activity. Both muscles had their highest activity during combined flexion and right rotation (45°). This activity was significantly ($p<0.0001$) different than pure exertion (0°, 90°) (see Figures 2-6, (bottom) and 2-7 (top), and Table 2.6).

The right EOB and left IOB had similar pattern activation and this pattern was the opposite of their bilateral muscles. The highest activity of these muscles were in combined left rotation and flexion (-45°), followed by pure flexion (0°), and combined right rotation and flexion (45°). There were no significant differences between pure flexion (0°) and flexion and right rotation (45°), Table 2.6.

The left RA showed maximum activity in pure flexion (0°), while the right RA showed higher activity in combined left rotation and flexion (-45°). The difference was not significant.

Both right and left ES showed the highest activity during combined extension and right rotation (135°), followed by pure extension (180°), and rotation (90°). The results of the The Tukey test suggests that the differences between pure extension and combined extension and right rotation were not statistically significant (Table 2.6).

Table 2.4. The outcome of univariate statistical test (ANOVA) of ten trunk muscles for the main effect of angle (5 levels), exertion (3 levels),and their interaction effect.

	Independent Variables		
	Exertion	Angle of moment	Exertion * Angle
Right EOB	171 (0.0001)	79 (0.0001)	4 0.0002
Left EOB	209 (0.0001)	101 (0.0001)	5 0.0001
Right IOB	146 (0.0001)	48 (0.0001)	2 0.03
Left IOB	110 (0.0001)	23 (0.0001)	3 0.012
Right RA	125 (0.0001)	156 (0.0001)	13 0.0001
Left RA	246 (0.0001)	325 (0.0001)	21 0.0001
Right ES	79 (0.0001)	139 (0.0001)	3 0.011
Left ES	148 (0.0001)	274 (0.0001)	5 0.001
Right LAT	181 (0.0001)	60 (0.0001)	4 0.004
Left LAT	67 (0.0001)	23 (0.0001)	3 0.01

Table 2.5 The P values of Tukey multiple comparison tests for main effect of trunk exertion (30, 60, %100 MVE) on NRMS-EMG of 5 bilateral trunk muscles.

	Left			Right		
	30-60	60-100	30-100	30-60	60-100	30-100
External Oblique	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
Internal Oblique	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
Rectus Abdominus	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
Erector Spinae	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
Latissmus Dorsi	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001

Table 2.6. The P values of Tukey multiple comparison tests for main effect of the angle (0° to 180°) of net resultant moments on the means of NRMS-EMG activity of right and left EOB, IOB and RA muscles.

	EOB		IOB		RA	
	Left	Right	Left	Right	Left	Right
0 - 45	0.0001	N.S.	N.S.	0.0166	N.S.	N.S.
0 - 90	0.0001	0.0001	0.0001	N.S.	0.0001	0.0001
0 - 135	0.0001	0.0001	0.0001	N.S.	0.0001	0.0001
0 - 180	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
45 - 90	0.0001	0.0001	0.0006	0.0207	0.0001	0.0001
45- 135	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
45-180	0.0001	0.0001	0.0001	0.0001	0.0001	0.0001
90-135	N.S.	N.S.	0.0055	0.0457	0.0322	N.S.
90-180	0.0001	0.0001	0.0035	0.0001	0.0001	0.0001
135-180	0.0001	0.000	N.S.	0.0001	0.0155	0.0054

Table 2.7. The P values of Tukey multiple comparison tests for main effect of the angle (0° to 180°) of net resultant moments on the means of NRMS-EMG activity of right and left ES and LAT muscles.

	ES		LAT	
	Left	Right	Left	Right
0 - 45	N.S.	0.0001	0.0071	0.0001
0 - 90	0.0001	0.0001	0.0001	0.0001
0 - 135	0.0001	0.0001	0.0034	0.0001
0 - 180	0.0001	0.0001	0.0001	0.0001
45 - 90	0.0001	0.0001	N.S.	N.S.
45- 135	0.0001	0.0001	0.0001	N.S.
45-180	0.0001	0.0001	0.0001	0.0001
90-135	0.0001	0.0001	0.0001	N.S.
90-180	0.0001	0.0001	0.0001	0.0001
135-180	N.S.	N.S.	N.S.	0.0001

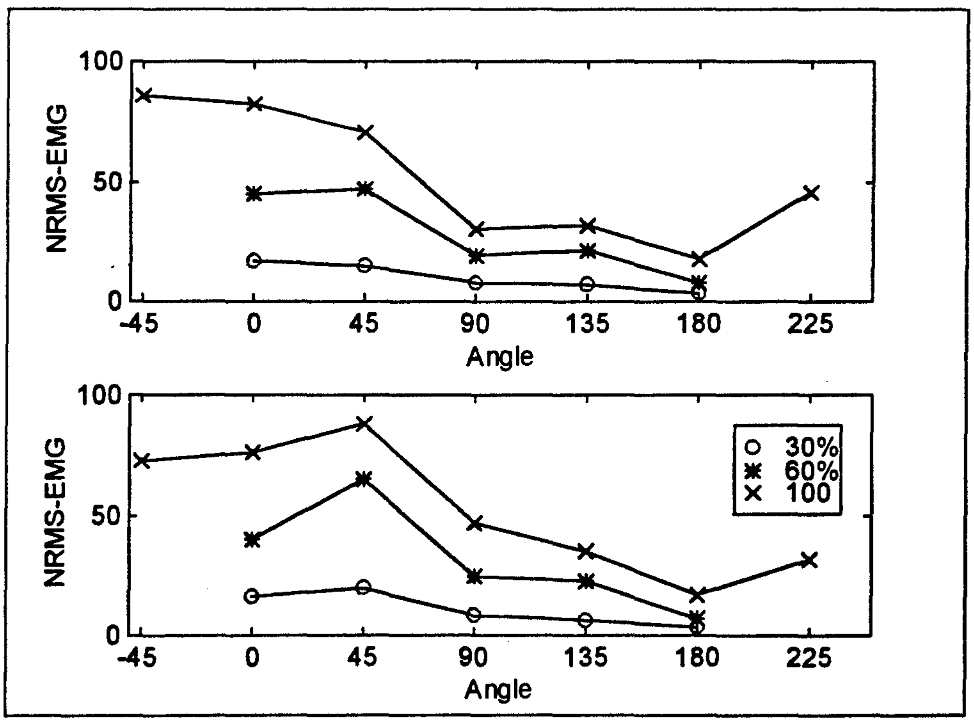

Figure 2.6. Mean NRMS-EMG of right (top) and left (bottom) external oblique muscle for different magnitude and angle of exertion.

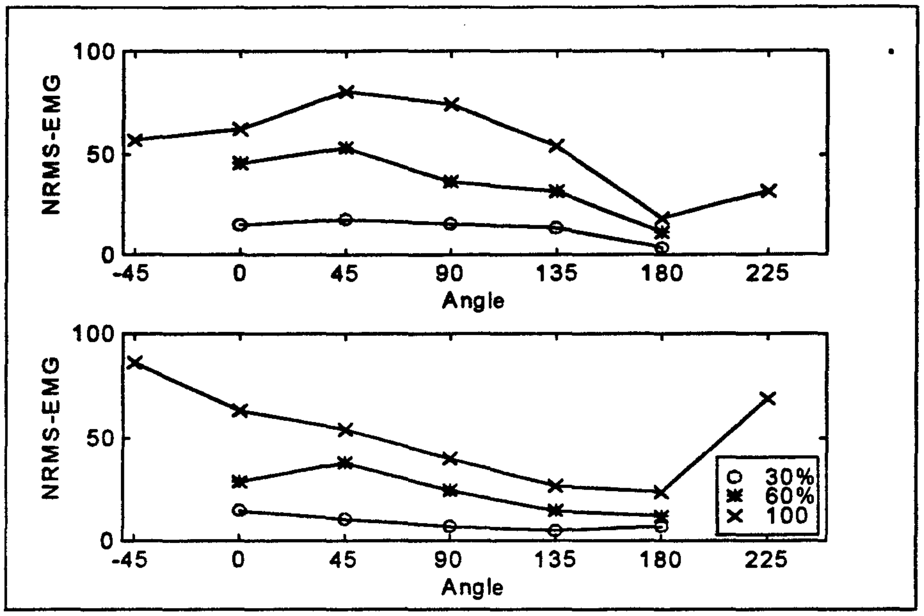

Figure 2.7. Mean NRMS-EMG of right (top) and left (bottom) internal oblique muscle for different magnitude and angle of exertion.

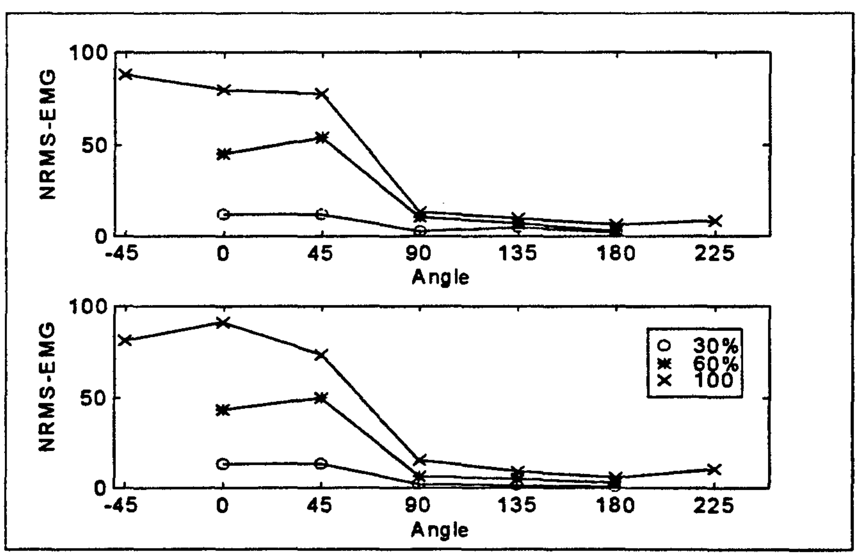

Figure 2.8. Mean NRMS-EMG of right (top) and left (bottom) rectus abdominis muscle for different magnitude and angle of exertion.

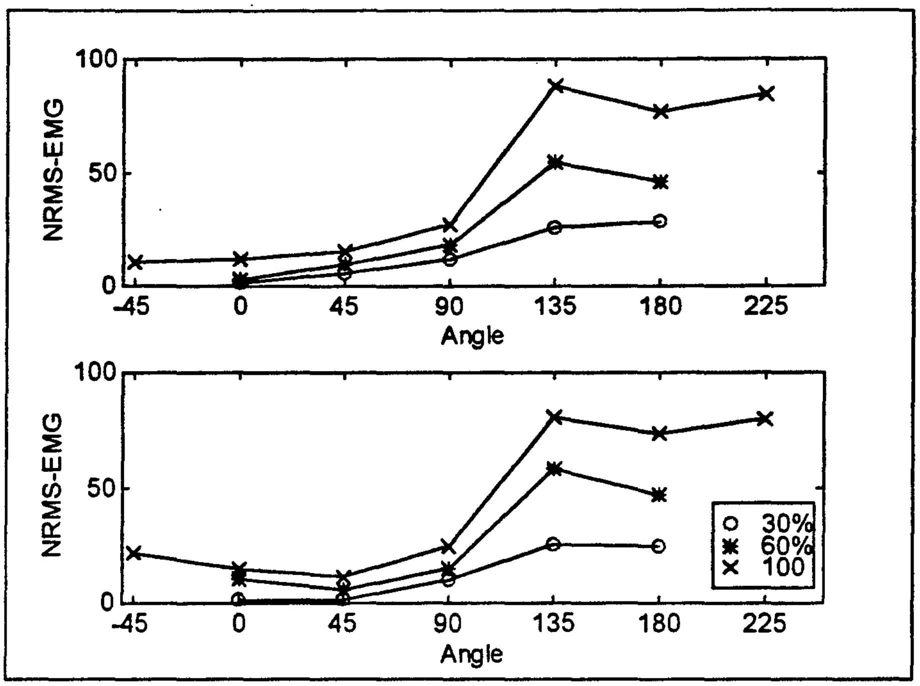

Figure 2.9. Mean NRMS-EMG of right (top) and left (bottom) erector spinae muscle for different magnitude and angle of exertion.

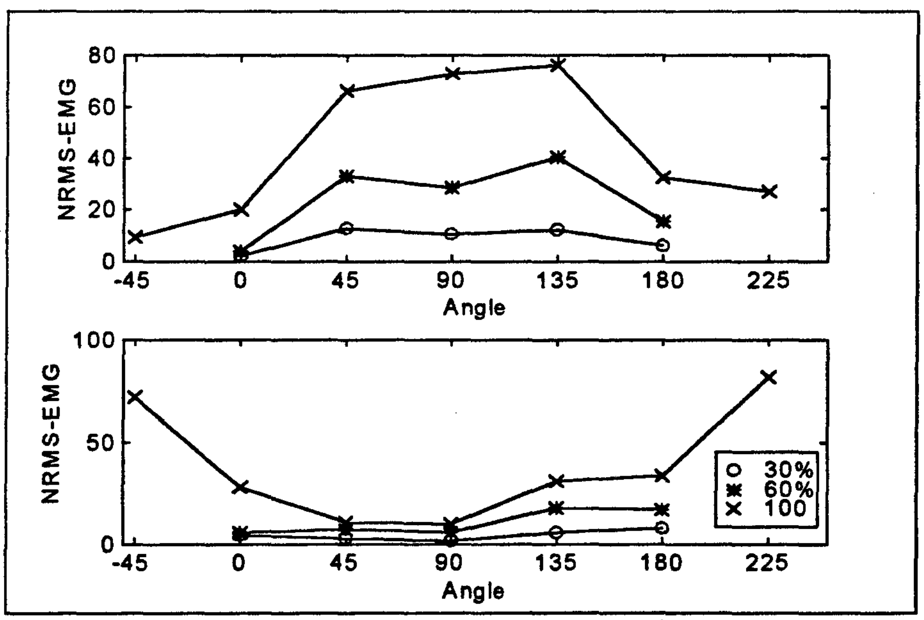

Figure 2.10. Mean NRMS-EMG of right (top) and left (bottom) latissimus dorsi muscle for different magnitude and angle of exertion

The right LAT muscle had the highest activity during combined right rotation and extension (135°), followed by pure right rotation (90°), and combined right rotation and flexion (45°). These three angles of exertion were not significantly different from each other, however they were all significantly different than the pure extension (180°), (Table 2.7). The left LAT muscle had the highest activity during combined left rotation and extension (225°), followed by the left rotation and flexion (-45°), (Appendix D , Table 2.5).

2.4.3 Torque Exertion

The mean and standard deviation (n=15) of the torques produced with respect to each plane of exertion is presented in Table 2.8. The mean of flexion torques produced during the 100% exertion was 120.0 (±37) Nm for pure flexion, 121.6 (±43) Nm for pure extension, and 53.7 (±19) Nm for pure axial rotation.

Table 2.8. The means and standard deviations of measured sagittal and transverse trunk moments (Nm) for three levels of exertion (30%, 60%, and 100% MVE) and different angle of net resultant moments during testing conditions (0° to 180°) and for the MVE trails (-45 and 225), (n=15).

	30 % MVE		60% MVE		100% MVE	
	Sagittal	Transverse	Sagittal	Transverse	Sagittal	Transverse
-45					105.6 (35)	-43.5 (16)
0	37.9 (12)	0.7 (1)	76.4 (24)	0.7 (1)	120.0 (37)	0.3 (3)
45	37.8 (12)	18.0 (7)	0.1 (1)	33.1 (12)	105.4 (33)	42.5 (15)
90	-0.1 (1)	17.9 (7)	-73.7 (26)	35.3 (13)	0.5 (2)	53.7 (19)
135	-37.2 (13)	17.8 (7)	-73.6 (26)	34.2 (15)	-110.0 (38)	45.9 (16)
180	-39.0 (13)	0.6 (1)	-73.8 (25)	1.7 (1)	-121.6 (43)	2.6 (4)
225					-97.0 (36)	-38.7 (17)

In order to evaluate the subject's performance with respect to the target torque, the control index based on Equation (2) was calculated. Table 2.9 presents the calculated value of this index for pure and combined testing conditions. The high value of this index represents a large distance between the actual exerted torque and the expected target torque. Table 2.11 illustrates that the subjects performed much less than the expected torques during all the combined tasks that required 100% of exertion.

The result of the statistical analysis reveals that the value of the control index was significantly ($p < 0.0001$) affected by the level of exertion (Table 2.10). The result of the Tukey test (Table 2.11) suggest that the value of this index during 100% exertion was significantly different than for 30 and 60 % exertion.

Table 2.9. The means and standard deviations of the control index (Equation (2)) for three levels of exertion (30%, 60%, and 100% MVE) and five angle (0° to 180°) of net resultant moments during testing, (n=15)..

	30 % MVE	60 % MVE	100 % MVE
0	1.25 (1.16)	6.72 (16.12)	8.75 (5.60)
45	0.98 (0.54)	4.10 (2.55)	28.53 (17.69)
90	0.68 (0.41)	1.69 (1.21)	7.41 (7.94)
135	1.13 (0.56)	3.09 (1.48)	22.33 (23.31)
180	1.14 (0.80)	2.65 (2.42)	5.62 (4.39)

Table 2.10. The outcome of univariate statistical test (ANOVA) of net resultant force for main effect of angle (5 levels), level of exertion (3 levels), and their interaction effect.

	Mean Square	F	P
Exertion	5387.67	45.50	0.0001
Angle	124.50	1.05	0.3087
Exertion X Angle	85.78	0.72	0.4881

Table 2.11. P value of Tukey multiple comparison tests on mean of level of exertion

	30% MVE	60% MVE
60	0.3686	
100	0.0001	0.0001

2.5 Discussion

A total of 15 subjects, 10 males and 5 females, with no history of back pain for the last six months participated in this study. The subjects had a mean age of 29.8 (±6.4) years. The mean MVE was 120 (±37) Nm for flexion, 121.6 (±43) Nm for extension, and 53.7 (±19) Nm for right rotation. The results of habitual physical activity questionnaire show that subject physical activities are within the range reported by Baecke et al.,(1985).

2.5.1 EMG Activity of Trunk Muscles

With respect to the primary hypothesis, the statistical analysis suggested that the mean of NRMS-EMG of the ten selected trunk muscles were significantly affected by the magnitude and orientation of the net resultant moment of the trunk. The significant interaction effect suggests that these factors have a combined effect on NRMS-EMG of trunk muscles, rather the individual factors affecting the activity muscle independently.

No other study similar to this study that has evaluated pure vs combined effect of trunk muscle exertions. However, results of other studies in this area support our findings. Several studies investigated the effect of upright twisting postures on maximum torsional exertion. These studies demonstrated that the maximum isometric torsion moment is a function of the twisting angle (Marras and Granata, 1995) and the direction of exertion. Furthermore, they reported a higher coactivation for all the trunk muscles during the twisting posture. The higher EMG activity of the trunk muscles was attributed to the changes in the length-tension curve of these muscles and the stability required during rotation (Marras and Granata, 1995; Pope et al., 1987).

Vink et al. (1992) collected EMG activity of the erector spinae and maximum isometric trunk extension torques during twenty three asymmetrical and symmetrical postures with a trunk flexion of 30° or 60° with respect to each axis. For asymmetrical exertion, they reported 30% reduction of force with respect to symmetric upright standing, and up to 40% with respect to symmetrical flexed posture. EMG activity of the RA muscle was higher during trunk rotation. The highest EMG activity of trunk muscles was reported during combined trunk flexion and rotation.

Only Lavender and his colleagues (1992) investigated the effect of asymmetrical exertion on symmetrical upright posture. Their study was similar to the present study with respect to quantifying the activity of trunk musculature as a function of the direction and magnitude of an external net resultant moment. However, Lavender et al. (1992) defined the net resultant moment as a combination of lateral and sagittal exertion; where as in our study, it is defined as a combination of sagittal and transverse exertion. Furthermore, the net resultant force in their study was 10 to 50 Nm with increments of 10 Nm (Lavender et al., 1992), and 20 and 40 Nm in another study (Lavender et al., 1993). In the present study, the magnitude of torque is defined as combination of 30%, 60%, and 100% MVE along each axis. Hence, the larger range of generated torques strengthens the generalizability of our findings.

In Lavender's (1992) study, the direction of net moment was changed from pure flexion to pure extension with 15 or 30 degree increments. Except for the left LAT muscle, all the trunk muscles were significantly affected by the interaction of the magnitude and angle of the net resultant forces.

In the present study, we attempted to decouple the EMG activity of the trunk muscle with respect to the activity required to generate moments along the sagittal and transverse axis. The rationale behind this approach is based on the anatomical and biomechanical observation of the trunk muscles during asymmetrical trunk exertion. The coactivation of the trunk muscles produces a net trunk moment along all three axes. In the case of planar flexion and extension, the RA muscles and ES muscles respectively are considered as primary movers. In both cases, these muscles are aligned within the plane of sagittal exertion. However, axial rotation is achieved by the net resultant of the coactivation of all trunk muscles and no single muscle is defined as the primary mover. The lack of primary muscles for axial rotation and the limitation of the number of trunk muscles raises questions concerning the allocation of muscles during combined exertion. One would expect that EMG activity of each trunk muscle during combined exertion should be equal to the sum of the EMG activity of that muscle during planar exertion.

Among anterior muscles, the mean of NRMS-EMG of the right IOB and left EOB muscles were significantly affected by the level and orientation of the net resultant moment and their interaction (Table 2.4). Independent of the level of exertion, Figures 2.6 (bottom) and 2.7 (top) and the results of the Tukey test (Table 2.6) show significantly higher levels of EMG activity for these muscles during combined exertion compared to pure flexion or pure right rotation. Table 2.6 shows that EMG activity of these muscles during combined exertion is less than the sum of NRMS-EMG of muscles during pure exertion along each axis. For example, during 100% MVE, mean the NRMS-EMG activity of the right IOB muscle was 62.47 (±22) during pure flexion, 74.29 (±26) during pure rotation, and 80.63 (±19) during combined exertion.

The mean of NRMS-EMG of the right IOB and left EOB muscles during combined extension and right rotation was significantly lower than during axial rotation and significantly higher than during pure extension. This study, as well other studies (Lavender et al., 1992; Lavender et al., 1993), assumed that the trunk performance is symmetrical for right and left rotation. Therefore, the left exertion was not examined in detail in this study and it was not included in the statistical analysis. However, Figures 2.6 (top) and 2.7 (bottom) illustrate that the left IOB and right EOB muscles have similar patterns as the right IOB and left EOB muscles in left rotation.

The right LAT muscle showed higher activity during combined right rotation and extension compared to pure extension along each axis individually (Figure 2.10). Similar to the pattern of the IOB and EOB muscles, the EMG of the activity muscles during 100% combined exertion was less than sum of the EMG activity of muscle during pure exertion. The highest muscle activity occurred during 100% extension and right rotation, followed by 100% pure rotation and rotation and flexion.

The RA and ES muscles are the primary movers in trunk flexion and extension. They are oriented along the sagittal plane and make a minimal contribution to the axial rotation torque. The left RA muscle showed the highest level of EMG activity during combined exertion; however, the activity level was not significantly different from pure flexion exertion. The right and left ES muscles had similar patterns of activity during combined extension and rotation compared to pure extension exertion.

In summary, all the trunk muscles showed higher activity during combined exertion. Although the amplitude of EMG activity was different, the pattern of muscle activity for the three levels of exertion was very similar. Compared to the ES and RA muscles, muscles that are oriented in an oblique direction (i.e., IOB, EOB, LAT), are taxed more during combined trunk exertion. Due to the long lever arm of these muscles, the higher activity of these muscles produce higher compression and shear forces on the spine and soft tissues, which lead to increased risk of low back injury. These results confirm epidemiological findings and suggest that the combination of high load and asymmetrical exertion is a risk factor for injury to the lumbar spine.

2.5.2 Moment Generated During Combined Exertion

Previous studies demonstrated a decrease in maximum back strength and maximum acceptable weight during asymmetrical lifting (Chapter I for more detail), (Vink et al., 1992). Garg and Badger (1986) reported a 7-22% decrease in the maximum acceptable weight as the angle of asymmetry increased during a lifting task. Vink et al.

(1992) reported a 30% reduction of trunk extension MVE during an upright twisting posture compared to symmetrical exertion.

Other studies demonstrated the trunk limitation in producing torsional MVE (Marras and Granta, 1995; McGill and Hoodles, 1990; Pope et al., 1987) or extension MVE (Kumar and Garang 1992; Vink et al., 1992) as a function of trunk upright twisting posture. These studies suggested that the reduction of MVE during asymmetrical exertion is due mainly to changes in muscle parameters, i.e., changes in the lever arm and centroid of the muscle. The present study is the only study that investigated the trunk muscles limitation in pure and combined sagittal and transverse exertion without the artifact of postural changes.

Independent of changes in muscle parameters, the trunk muscles are limited in performing asymmetrical exertion. Compared to pure sagittal exertion, the sagittal combined torque was reduced by 12% in flexion and by 10% in extension (Table 2.9). Compared to pure transverse exertion, the combined torsional torque was reduced by 15% in combined flexion torque and 20% in combined extension torque (Table 2.9). Furthermore, Table 2.9 shows similarity between right and left trunk performance with respect to the maximum sagittal and transverse torque exertion during a combined task.

The torque generated during MVE is thought to be an indication of the maximal capacity of an individual for handling an external load. If the trunk is thought of as a linear mechanical system, then a combined exertion would be the vectorial sum of the maximum exertions in the pure planes. The significant difference between the actual the exerted moment and the expected torque suggests that the vectorial sum of the planar exertion significantly overestimates the trunk performance during the combined exertion. Figure 2.11 demonstrates the capability of the subjects in performing combined trunk exertion. The figure presents actual torque performed by the subjects during pure and combined 100% MVE. The torque values are normalized to pure planar exertion. Figure 2.11 shows that subjects are not capable of producing 100% combined MVE exertions. Furthermore, compared to planar exertion, there is high variability among the subjects in performing the combined task.

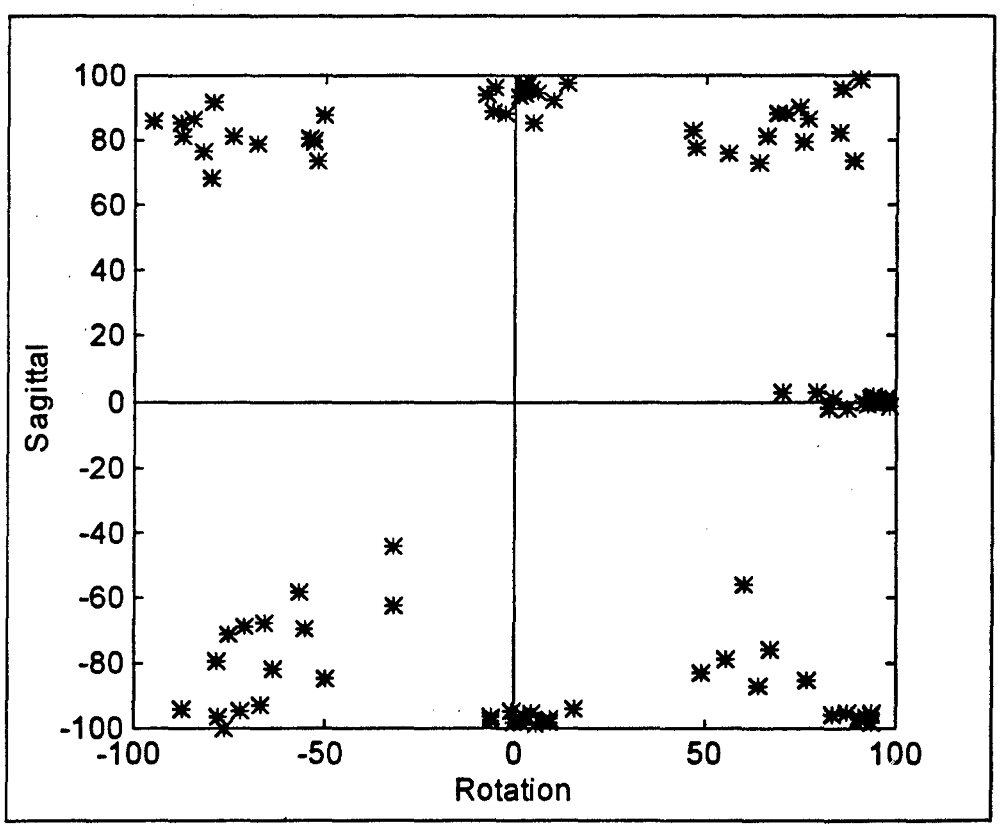

Figure 2.11: The sagittal and transverse torques produced by 15 subjects during 100% pure and combined exertion. The subject data is normalized to 100% of MVE.

It is generally believed that coupled trunk exertion (asymmetric push, pull, and lifting) is hazardous to the musculoskeletal system. In light of the results of this study, all trunk muscles were significantly affected by the level and angle of exertion. A higher level of coactivity leads to a higher internal loading of the spine, in addition to larger strain and stress in the annulus fibers and facet joints, and hence to a higher risk of injury.

2.5.3 Implication of Study

Measurement of trunk strength is frequently used in clinical evaluation of the LBP patients, as replacement test for certain jobs, and research studies. Maximum voluntary exertion is the manifestation of the capability of voluntary recruiting the maximum resources available to the neuromuscular system. Therefore, the maximum moment exerted on external devices, i.e., dynamometers and EMG signal collected from a muscle during MVE, can serve as a standard unit to gauge other trunk performances. For instance, the moment and EMG signal measured under MVE is used as a means to evaluate the overall effects of any clinical or experimental intervention (i.e., exercise training, fatigue, immobilization).

The implicit rationale for using the MVE as a unit of measurement is the reliability and validity of MVE in recruiting the maximum resources of the neuromuscular system (Enoka and Fuglevand, 1993). This reliability and validity should be reviewed carefully in light of our knowledge of human performance and motor control.

The MVE can be influenced by several physiological and biomechanical factors (Enoka and Fuglevand 1993). For instance, the MVE depends on joint angle, direction of testing, model of testing, and individual motivation. This suggests that the reliability and validity of MVE affect by the testing protocol. The standardization of testing protocol will limit the influence of some of these variables and minimize the strategies that the neuromuscular system uses to execute the task.

The results of this study raise some concern with respect to the measurement of mechanical and EMG data during MVE. Commonly, the protocols for measuring trunk strength consist of quantification of trunk maximum voluntary exertions and planar exertion along each trunk axis. The torque generated during MVE is thought to be an indication of the maximal capability of an individual for handling an external load (Ayoub and Mital, 1989). It is questionable how the information derived from pure plane exertions can be used to predict the capabilities of an individual in situations that involve combined exertions. The results of the present study indicate that it is incorrect to predict an individual's ability to perform high-level combined exertion based on planar trunk exertion performance. An estimation of individual performance based on a vectorial sum of planar exertion will significantly over-estimate the capability of performing combined trunk exertion. Industrial jobs often require individuals to have high capability and controllability during asymmetric lifts, i.e., lifting heavy objects and placing them at special area on shelf. In the ergonomic design of these jobs, the results of this study suggests that both capability and controllability of the subjects should be considered. Furthermore, higher subject variability can be seen in performing asymmetrical tasks.

To enhance the biomechanical interpretation, the EMG signal of a muscle is often normalized to the range of the highest and lowest activity of the muscles observed in the experiment. Generally, the highest activity of a muscle is taken from the isometric exertion along each axis of the trunk. The statistical analysis reveals that the NRMS-EMG activity of the LAT, IOB, and EOB muscles is significantly higher during the combined than the pure exertion. This implies that the activity of each measured muscle during combined exertion is a better estimate of the maximal activity of the muscle. Therefore, combined exertion value should be used for the normalization of the muscle.

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Chapter III

EMG-Driven Model of Trunk

3.1. Review of the Literature

Epidemiological data suggests a relation between exposure to heavy load and the incidence of low back disorder. However, the exact nature of this relationship is not known. Several methods have been suggested by biomechanicians and ergonomists to identify the risk of low back injury based on information gathered from tasks and individual methods of performance. The identification of a hazardous task can be based on a simple ergonomic checklist (Halpern, 1992, Mital, 1996) or on advanced three-dimensional biomechanical models. The criteria for the selection of methods and tools used for identifying hazardous tasks is mainly dependent on the accuracy and reliability required to estimate the risk of injury.

Ergonomic checklists (Mital, 1996) inspect the existence of risk factors by applying general biomechanical and ergonomical concepts. The content and underlying theoretical framework is usually derived from expert opinion. Therefore, a ergonomic checklist can provide a simple but less accurate estimation of job or task effort and the risk of injury.

Biomechanical models provide a more accurate estimation of load exposure and thereby risk of injury by estimating the net internal forces. Various static (Chaffin, 1969; Chaffin and Erig, 1991; Hans et al., 1991; Ladin et al., 1991; McGill, 1991; Schultz et al., 1987) and dynamic (Freivalds et al., 1984; Marras and Granata, 1995; Marras and Sommerich, 1991) models are proposed to predict the net force on the spine. The rationale for the selection of an adequate biomechanical model depends on the complexity of the task, the reliability and validity required for output information, the simplicity and reliability of gathering input information, the sensitivity of the model to input information, and the acceptability of the assumptions made by the model.

The two-dimensional static models are based on equivalent muscles so that the unknown muscle forces will be equal to the number of equations available that satisfy the equilibrium conditions. These models calculate the compressive and shear forces on the spine, mainly at the L3-L4 level (Chaffin, 1969; Chaffin and Erig; 1991). These models do not consider individual muscle forces in the prediction of the load on the spinal structure. Therefore, they are not adequate for estimating the internal load of the trunk during asymmetrical exertion where the internal force increases due to high coactivation of muscles while the external load is fixed. These models offer a less complex calculation of the estimating load distribution on the spine structure. However, the reliability of estimated force decreases as the coactivation of trunk muscles increases during task performance.

The expansion of the two-dimensional models leads to indeterminate models and redundant systems. For these models, a solution is obtained by defining system constraints and formulating an objective function that can be utilized by the optimization technique. For formulation of the system equation for the trunk, the external moment of the system needs to be balanced by the internal moments. The internal moments are based on the description of muscle forces and their moment arms. For the optimization model, the constraints of the system are based on the maximum muscle exertion and the biomechanical threshold of the soft tissues. The objective function should be derived based on the biomechanical principles and neuromuscular criteria of the human performance. However, these criteria are poorly known. Most of the proposed objective functions are based on minimizing the compression and shear forces, muscle stresses, or by some combination of these parameters (Brand et al., 1986; Crowninshield and Brand, 1981; Genaidy and Houshyar, 1989; Han et al., 1991; Parnianpour, 1991; Schultz et al., 1983). Depending on the objective function, the estimated spinal forces can vary significantly. The reliability of these models depends solely on the reliability of the objective function and the anatomical information used in the model.

An EMG-driven model estimates the compressive and shear forces on the spine by calculating the resultant forces of all the trunk muscles. The individual muscle forces are estimated from EMG activity. These models are found to be less adequate for predicting axial rotation (Marras and Granata, 1995; McGill, 1991), but they produce reasonable results for tasks within the sagittal plane (McGill and Norman, 1987).

The reliability of estimated force and moments of the trunk by EMG-driven models is dependent upon the reliability and accuracy of the anatomical parameters of the trunk muscles (cross-sectional area, lever arm, and line of action), EMG signals, and the assumptions made with respect to the EMG-force relationship.

A majority of the previous models assumed a linear EMG-force relationship in the formulation of these models. Both evidence of linear (Lippold, 1952; Woods and Bigland-Ritchie, 1983) and nonlinear (Komi and Buskirk, 1970, Zuniga and Simons, 1969; Lawrence and De Luca, 1983) EMG-relationships exist in the literature. Furthermore, they assume that the linear relationship is similar for all trunk muscles. The purpose of this study is to: a) present a general EMG-driven model where the individual EMG-force relation can be predicted for each muscle, b) compare the results of the EMG-driven models for three selected assumptions with respect to the EMG-force relationship, and c) test for the hypothesis that the results of EMG-driven models will be affected significantly by the assumptions made with respect to the EMG-force relationship of the trunk muscles.

3.2. Method

Four male and two female subjects between the ages of 20-40 years with no history of low back pain for the last six months volunteered to participate in this study. All subjects were screened for neurological or physiological conditions that would make them unsuitable for this study. Screening was done using a questionnaire and interview.

3.2.1. Instruments

A multichannel data acquisition system was developed in-house to simultaneously collect 16 channel EMG signals and 16 channel mechanical data from B200 Isostation (Isotechnologies Inc., Hillsborough, NC). The Labview software (National Instrument, Austin, TX) was used to collect data from data acquisition system via A/D board (ATMIO-64F, National Instrument, TX) at a sampling rate of 1000 Hz. The software was programmed to provide real-time visual feedback regarding the trunk torques measured by the B200 Isostation. More details and technical information regarding the instruments and experimental setup are provided in Chapters II and IV.

3.2.2. EMG Electrodes

Both surface and wire electrodes were used to detect myoelectric activity of the trunk muscles. The surface electrodes were bipolar, pre-gelled electrodes (Classic Medical, Muskego, WI). The bipolar wire electrodes were made in-house based on methods described by Basmajian and De Luca (1985). For the best mechanical strength and stiffness, a wire with 10% Iridium- 90% platinum alloy (Medwire, Mount Vernon, NY) was used. The wire was Teflon coated with a full coated diameter of 0.0014 inches. To make the wire electrode, the 25 cm Teflon coated wire was looped and passed through cannula of 26 G X 3.5" hypodermic spinal tape needle (Baxter Health Care Co., Edison, NJ). Then the loop was cut into two pieces of wire. About 2 mm of each wire at the distal end of the needle was exposed to flame in order to burn the Teflon isolation. The electrodes were sterilized in the Hospital for Joint Diseases before use.

3.2.3. Procedure

All subjects were asked to fill out the activity questionnaire (Baecke et al., 1982), demographic and medical questionnaire (see Appendices A and B), and sign the consent form (see Appendix E). The demographic and medical questionnaires were reviewed to ensure that the subject satisfied the inclusion criteria. The subject was asked to wear comfortable sport wear and shoes during testing.

Step 1. Electrode placement

Both wire and surface electrodes were used to collect EMG signals from five bilateral trunk muscles. Wire electrode was used for the right and left internal oblique (IOB) muscles and surface electrode was used for external oblique (EOB), erector spinae (ES), rectus abdominis (RA),and latissimus dorsi (LAT) left and right muscles. In Chapter II, the details of muscle landmarks used for surface electrode placements are reported. To insure the safety and accuracy of wire electrode placement, the wire electrodes were placed using needle biopsy procedures under CT-scan by a board certified radiologist. Upon completion of the questionnaire, the subject was scheduled for a CT-scan (CT 9800 HiLight) at the Radiology Department, Hospital for Joint Diseases, New York University Medical Center.

Each subject was placed in a supine position on the CT-scan table. Before and during the procedure, the subject was instructed to relax and maintain his/her position. The L3/L4 level was verified from the sagittal plane image of the trunk. Two 10 mm slice thickness images were obtained at the L3/L4 level in order to optimally localize the internal oblique muscles in the transverse plane. This location was marked on the subject's skin with ink, and radiopaque beads were placed at 1 cm intervals bilaterally in this transverse plane above the expected location of the internal oblique muscles. A CT image showing the marker beads on the skin allowed a selection of the correct sagittal plane and therefore the best point on the skin surface to advance the electrodes. At this level, markers were placed on the skin. A CT-scan was then taken at the L3/L4 level in the transverse plane. Based on the marker placement and location of the internal oblique muscle, the site of insertion of the wire electrode was selected.

Following sterile preparation, with CT scan guidance, the electrodes were slowly placed into the internal oblique muscle. No local anesthetic was necessary. Because the electrodes could not be pulled back and re-advanced, it was essential to slowly advance each needle with repeated CT imaging to check the position. In most cases, the electrode reached the mid-third of cross-sectional area of IOB muscle within a three-step advancement of the electrode. Once the electrode reached the mid-third of cross-sectional area of IOB muscle, the spinal needle was removed and the wires were inspected and taped to the body.

Upon the completion of the wire electrode placement, the subject was then brought to the Muscle Recruitment Laboratory in the hospital. The site of surface electrode placement for four bilateral muscles, EOB, RA, ES at L3/L4 and at L1-T12 level for LAT, was prepared, based on anatomical considerations, palpation of muscles during contraction, and the recommendations of previous studies (Gabriel, 1992; Ross et al., 1993; Schmitz, 1992; Tan et al., 1993). Further details with regard to the electrode placement sites are provided in Chapter II.

Step 2. Maximum Voluntary Exertion (MVE)

The participant was strapped to the B200 Isostation in an upright standing posture according to the method described in the B200 User's Manual (Isotechnologies, Inc.,

1988) and the procedure described in previous studies (Gabriel, 1992; Ross et al., 1993; Schmitz, 1992; Tan et al., 1993). The subject performed two trials of MVE of the trunk in an upright standing posture in four random directions: flexion, extension, and right and left axial rotation. The secondary axes were unlocked during all the trials. The subject was asked to reach to his/her MVE and hold it for 3 seconds. Two minutes of rest separated each test to avoid fatigue.

Step 3. Pure and combined exertion

Based on the MVE values, the computer randomly selected a test from combinations of 0, 30, 60, and 100% of MVE for flexion, extension, and rotation of the trunk. To simulate pure and combined exertions, visual feedback was given to the subjects. A monitor visually displayed trunk torque output and target torque in the plane of sagittal and axial torque. The lateral axis was unlocked, hence provided no resistance along this axis. The computer provided an audio warning if a subject exceeded ±5 degree of lateral bending. A target torque was set at 0, 30, 60, 100% of MVE exertion along one or combination of two axes of the trunk. Each subject was instructed to match their output torque to the target torque for at least three seconds. To reduce the number of testing conditions and the risk of fatigue, only rotation to the right side is considered in this study (see Figure 2.4 for test condition used in this study). No load condition, zero condition along each axis, was repeated more than three times. However, these tests were part of the randomization process. The zero condition served as a means to calculate the baseline EMG for the normalization of the data.

Before each test, the feedback screen was displayed on the terminal and the subject was asked to practice reaching the target area. Based on the subject's performance, verbal feedback was provided. The practice trial was continued until the subject felt comfortable with the task and indicated that he/she was ready to start the test.

3.2.4. Data Analysis

3.2.4.1. Data reduction

Based on Equation (1), three seconds of each trial was selected in which the subject performed closest to the target torque. The close distance was defined by the low value of the control index as it was calculated by the following formula:

$$\text{Control Index} = \frac{\sqrt{\left(Sag_{Torque} - Sag_{Tag}\right)^2 + \left(Tran_{Torque} - Tran_{Tag}\right)^2}}{\sqrt{\left(Sag_{Tag}\right)^2 + \left(Tran_{Tag}\right)^2}} \tag{1}$$

Where :

Sag_{Torque} and $Tran_{Torque}$ = The generated sagittal (Sag) and transverse (Tran) torque

Sag_{Tag} and $Tran_{Tag}$ = The sagittal (Sag) and transverse (Tran) target torque

For all ten trunk muscles, the mean of the normalized Root Mean Square of the EMG signal (NRMS-EMG) was calculated for a selected 3 second window. Based on previous recommendations (Redfern, 1992), the time constant of 50 ms was used for calculation of the RMS. As suggested by Lavender et al. (1992) and Seroussi and Pope (1987), the EMG signal of each muscle was normalized to the maximum and the rest value of the muscle. In this study, the highest mean NRMS-EMG values among all the trials were selected for the normalization procedure.

3.2.4.2. CT-SCAN

The CT-scan images were down loaded from the CT-scan unit to optical disks. At the Radiology Department of the School of Medicine, New York University, the images were converted to 512x512 binary image files. Sigma Scan Pro image processing software (Jandel Scientific, San Rafael, CA) was used to process the CT-scan images. From the CT-scan images, the width and depth of trunk, cross-sectional area, and distance of centroid of muscle from the center of the disk were measured. The sagittal view of the spine was inspected to insure that the angle of L3/L4 vertebra with respect to the horizontal line was about 90°, and that the values measured from these images did not need any correction, as suggested by McGill et al. (1993).

3.2.4.3. Calculation of Muscle Forces

The EMG-driven model was used to predict trunk moments and spinal forces. The EMG-driven model is based on the assumption that the tensile force generated by a muscle is related to the normalized EMG and cross-sectional area of a muscle. There are controversial reports with respect to the characterization of this relationship as linear (Lippold, 1952; Woods and Bigland-Ritchie, 1983) and nonlinear (Komi and Buskirk, 1970, Lawrence and De Luca, 1983; Zuniga and Simons, 1969). In this study as well as others (Granata, 1993), a linear relation between tensile force and EMG has been employed, based on several factors: 1) the nonlinear relationship for the mid-range of the force can be approximated as linear (Granata, 1993; Solomonow et al., 1986); 2) when the coactivation of muscles was accounted for in a model, based on Hof and Van Den Berg (1977), the EMG-force relationship was linear; and 3) for simplicity of formulation, the relationship can be assumed to linear and may be expanded to be nonlinear in the future.

If the relation between the myoelectrical activity of muscle and tensile force is assumed to be linear, then the tensile force (F_i) of muscle i is:

$$F_i = G_i \times (A_i \times NEMG_i) + b_i \qquad (2)$$

where:

G_i = Gain of ith muscle
$NEMG_i$ = Mean of normalized RMG-EMG of ith muscle
A_i = Cross-sectional area of ith muscle
b_i = intercept of line

Since the muscle force cannot be measured directly, G_i can be calculated from moment equations, where the measured moment (M) for the j axis is:

$$M_j = \sum_{i=1}^{n} \left(G_i \times (A_i \times NEMG_i) + b_i\right) \times L_{ij} \qquad (3)$$

$$M_j = \sum_{i=1}^{n} \left(G_i \times A_i \times NEMG_i \times L_{ij}\right) + \left(b_i \times L_{ij}\right) \qquad (4)$$

where L_{ij} is the lever arm of muscle i in j plane, and n is the number of muscles, which in this study is equal to ten.

In this study, three models have been formulated based on the assumptions that can be made with respect to G_i and b_i.

<u>Model I</u>: Based on previous studies, two assumptions can be made:

a) The gain of all muscles is equal to a constant G value:

$$G = G_1 = G_2 = \ \dots\ = G_{10}$$

b) The intercept of all muscles is equal to zero:

$$b_1 = b_2 = \ldots = b_{10} = 0$$

Based on these two assumptions, Equation (4) can be simplified as:

$$M_j = G \times \left(\sum_{i=1}^{10} A_i \times NEMG_i \times L_{ij} \right) \tag{5}$$

Model II: Similar to Model I, this model assumes that the gain of all trunk muscles is equal to a constant G value. Furthermore, it assumes that b_i of all the muscles is equal to each other:

$$b = b_1 = b_2 = \ldots = b_{10}$$

Based on this assumption, Equation (4) can be simplified as:

$$M_j = \sum_{i=1}^{10} \left(G \times A_i \times NEMG_i \times L_{ij} \right) + \left(b \times L_{ij} \right) \tag{6}$$

Model III: Specific conditions can be assumed; that the trunk muscles can be grouped together based on their functional (i.e., flexors vs. extensors) or physiological (i.e., fiber types) properties. For instance, in Model III, we assume that:

$$M_j = \sum_{p=1}^{m} M_{jp} \tag{7}$$

where :

$$M_{jp} = \sum_{i=1}^{n_p} \left(G_p \times A_i \times NEMG_{i_p} \times L_{ij_p}\right) + \left(b_p \times L_{ij_p}\right) \quad (8)$$

m is the number of grouped muscles and it can take any value from one to ten. Equations (4) and (7) present cases in which m is assumed to be one and ten respectively. In this model, we assume two groups of muscles: 1) extensor muscles that include the right and left LAT and ES muscles , and 2) flexor muscles that include the right and left RA and IOB and EOB muscles. For calculation of muscle force in these three models, the line of action of the muscle was used as reported by Nussbaum et al. (1995).

3.2.4.4.Ridge Regression

A detailed explanation of the ridge regression method and its derivation is presented by Neter et al. (1983). The following is brief summary of this method. For ordinary least squares,

$$y = bx \quad (9)$$

Neter et al., (1983) showed that the regression coefficient can be calculated by:

$$b = \left(r_{xx}\right)^{-1} r_{yx}$$

Where r_{xx} is a correlation matrix of all the independent variables, and r_{yx} is the vector containing the coefficients of simple correlation between the dependent variable and each of the independent variables.

The ridge regression estimates are obtained by (Neter et al., 1983):

$$b^r = \left(r_{xx} + CI\right)^{-1} r_{yx} \tag{10}$$

where C is a bias constant ($C \geq 0$), b^r is the vector of standardized ridge regression coefficient , and I is identity matrix.

The value of C indicates the amount of bias in the estimators. The value of C can range from zero to one. When C is equal to zero, Equation (10) is the ordinary least square regression. Neter et al. (1983) indicate that as C increases from zero to one, the total mean squared error of the ridge regression estimator b^r increases, while the variance component decreases.

Generally, the ridge trace is used to estimate the C value. The ridge trace is a plot of the b^r value (Equation (10)), when C takes a value between the range of zero to one. The ridge trace usually has a large fluctuation for the slight change of the C value from zero, and then tends to be stabilized for higher C values. By the visual inspection of the trace, one should select the smallest value of C where all the traces are relatively stable. In this study, to reduce the risk of error due to visual inspection of the ridge trace, we selected the value of C=0.01 for all the models. This value was selected since there is no criteria available for proper selection of C value any pre-select method might produce systematic error.

To solve Equation (9) by means of ridge regression, the data was arranged for each axis as follows:

$$M_j = \begin{bmatrix} a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \end{bmatrix} \begin{bmatrix} G_1 \\ \vdots \\ G_n \\ b_1 \\ \vdots \\ b_n \end{bmatrix} \tag{11}$$

where:

$a_{mn} = A_{mn} \times NEMG_{mn} \times L_{mn}$ for n^{th} muscle in trial m.
ℓ_{mn} = Lever arm of the n^{th} muscle in trial m
G_n = the gain of n^{th} muscle
b_n = the intercept of n^{th} muscle

To calculated the gain and intercept based on three axes, the Equation (11) should be modified to include the data of all three axis in ridge regression model. Equation (12) shows the expansion of Equation (11) where the data of all three axis were appended to a large matrix.

$$\begin{bmatrix} M_{sagittal} \\ \\ M_{Transverse} \\ \\ M_{Lateral} \end{bmatrix} = \begin{bmatrix} a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \\ a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \\ a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \end{bmatrix} \begin{bmatrix} G_1 \\ \vdots \\ G_n \\ b_1 \\ \vdots \\ b_n \end{bmatrix} \tag{12}$$

3.2.5. Statistical Analysis

For each subject, the mean of absolute error (the difference between the predicted and the actual moment produced) of trunk moments, compression, and shear forces were calculated. Descriptive statistics (the mean and standard deviation) were computed for demographic data, absolute error of moments, compression, and shear forces at the level of L3/L4 of the spine.

The statistical procedure MANOVA of repeated measure design (SAS Institute Inc., 1985) was used to determine the effect of the method of calculation (models) used on the results of the EMG-driven model of trunk. If the overall multivariate analysis was significant, a univariate analysis (ANOVA) was performed. The Tukey test was performed on the levels of the model if the effect of the univariate test was statistically significant. In this study, statistical significant level was set at $P<0.05$.

3.3. Results

The CT-scan and EMG data of 6 subjects (4 males and 2 females) was used in the analysis. Mean age, weight, and height were 27.3 (±5.4) years, 68.0 (±6.9) Kg, and 169 (±8.8) cm, respectively. The scores for index of work, sport, and leisure were 2.40 (±0.66), 2.95 (± 1.68), and 3.25 (±0.57) respectively. Table 3.1 summarizes the mean and standard deviation of the muscle parameters of the six subjects based on values measured from the CT-scan each individual subject. Additional data from ten subjects were collected but not included in this data analysis due to different technical reasons related to either CT-scan data (i.e., low resolution for separating the muscle or the loss of calibration information of the image), or missing EMG signal data (i.e., resulting from the loss of the wire or surface electrode signal), (Chapter II for detailed information).

For each subject, the mean of absolute error across all the trials was calculated. Table 3.2 shows the mean of absolute error for each plane, and the mean of spinal forces for all subjects. The sagittal moment error was 26.0 (± 4.7) Nm for Model I, and was reduced to 22.3 (± 9.2) Nm and 17.5 (± 5.7) Nm for Models II and III respectively. Similar level of errors were observed for the results of the three models with respect error in predication of transverse and lateral moments.

With respect to spinal forces, the three models differed in their ability to predict compression forces and lateral shear forces. The results of Model II predicted the highest lateral shear and anterior-posterior (AP) shear and compression forces compared to the other two models. Model III resulted in the lowest compression force with a mean of 1614.02 (± 796.5) N.

The result of the multivariate analysis of variance indicated a significant ($p<0.001$) effect of the models on the results of the EMG-driven model. The repeated ANOVA was performed to investigate the effect of the model on the mean of the predicted moments and spinal forces. Table 3.2 shows the summary of the results of the ANOVA procedure.

Table 3.3 presents the results of the post-hoc Tukey test for the main effect of the model on the mean of sagittal error, lateral shear and compression forces. Model I had significantly higher ($p <0.0049$) sagittal error compared to Model III. With respect to

Table 3.1. Mean and standard deviation of the cross-sectional area and the lever arm (the distance of the centroid of the muscle to the center of disc) for ten selected trunk muscles at the L3/L4 level of the spine.

	Lateral Lever Arm (mm)	Anterior-posterior Lever Arm (mm)	Cross-Sectional Area (mm^2)
Left ES	37.32 (3.81)	59.54 (4.59)	2506.79 (443.61)
Right ES	35.49 (7.94)	59.17 (5.63)	2540.37 (400.20)
Left RA	41.91 (10.12)	74.33 (13.02)	855.81 (201.36)
Right RA	44.93 (3.0)	75.02 (15.81)	884.83 (168.17)
Left IOB	108.42 (11.13)	17.81 (8.06)	1458.91 (406.86)
Right IOB	110.78 (11.00)	19.32 (11.32)	1317.51 (335.85)
Left EOB	121.2 (10.83)	15.2 (10.06)	1335.25 (357.79)
Right EOB	122.29 (10.15)	16.19 (12.02)	1320.09 (239.49)
Left LAT	102.56 (11.61)	39.93 (11.83)	193.62 (54.46)
Right LAT	102.53 (10.01)	39.62 (14.99)	215.63 (52.97)

lateral shear force, Model I predicted significantly (p <0.0052) lower lateral shear force. Significantly (p<0.02) higher compression force was predicted by Model II in comparison to the other two models.

Table 3.2. The mean and standard deviation for results of EMG-driven models and univarite analysis of variance (ANOVA) for the mean of the absolute predicted moments and compression and shear forces.

	Model I	Model II	Model III	F and P Value
Sagittal Error (Nm)	26.0 (4.7)	22.3 (9.2)	17.5 (5.7)	6.51 (0.015)
Transverse Error (Nm)	19.9 (7.5)	18.6 (8.8)	19.1 (8.1)	NS
Lateral Error (Nm)	14.1 (6.2)	15.6 (5.0)	13.9 (4.1)	NS
Lateral Shear (N)	66.7 (25.2)	72.0 (25.5)	72.4 (25.9)	7.90 0.008)
AP Shear (N)	448.7 (148.2)	495.7 (181.4)	424.8 (96.4)	NS
Compression (N)	1631.4 (718.0)	3961.3 (2597.9)	1614.02 (796.5)	4.79 0.035

Table 3.3. The P value of the Tukey multiple comparison test for main effect of Model on the means of EMG-driven model of trunk.

	Model I Vs II	Model I Vs III	Model II Vs III
Sagittal Error	-	1>3 0.0049	-
Lateral Shear	1 < 2 0.0078	1 > 3 0.0052	-
Compression	1 < 2 0.0234	-	2 > 3 0.0226

Table 3.4 Predicted gain (N/m^2) and intercept (Nm) based on three EMG-driven models for each subject (ID) and the mean and standard deviation (STD) for all six subjects.

	Model 1	Model II		Model III			
ID	Gain	Gain	Intercept	Flexor Gain	Extensor Gain	Flexor Intercept	Extensor Intercept
1	0.55	0.55	0.1116	0.59	0.53	-0.0826	-0.2322
2	0.28	0.36	0.8658	0.33	0.40	0.2259	0.0453
3	0.44	0.51	-0.4224	0.58	0.45	-0.1014	-0.1428
4	0.97	1.04	-0.8132	1.09	1.05	-0.5357	-0.6668
5	0.57	0.60	-0.3334	0.69	0.56	-0.0788	-0.0823
6	0.43	0.43	0.2929	0.63	0.41	-0.0082	-0.0494
Mean	0.54	0.58	-0.04987	0.65	0.57	-0.09683	-0.1880
STD	0.23	0.24	0.5973	0.25	0.24	0.2470	0.2522

Table 3.4 demonstrates the estimated gain and intercept of the EMG-force relationship as predicted by the ridge regression model. The lowest gain across all three models was 0.28 N/m^2 and the highest was 1.09 N/m^2. The mean of predicted gain ranged from 0.54 to 0.65 N/m^2. Subject 4 had the highest gain value across all three models. The intercept estimated for Models II and III ranged from -0.813 Nm to 0.865 Nm across all subjects.

3.4. Discussion

The statistical analysis suggests that the results of the EMG-driven models are significantly affected by the methods of muscle force calculation. The three mathematical formulations of force in this study stem from the possible theoretical assumptions made with respect to the EMG-force relationship.

Many studies have been performed to determine the relationship between surface EMG and the force generated by the muscle during various type of contraction. The

EMG-force relationship has been shown to be linear (Lippold, 1952; Woods and Bigland-Ritchie, 1983) and nonlinear (Komi and Buskirk, 1970; Lawrence and De Luca, 1983; Zuniga and Simons, 1969). The discrepancy with respect to the reported relation is mainly due to a variation of investigated muscles and the method used by the investigator (Perry and Bekey, 1981).

In addition to methodological differences in collecting force and EMG data , Hof and Van Den Berg (1977) demonstrated that the EMG of a muscle is linearly related to muscle force, but coactivation muscles generates a nonlinear relation between the EMG and the joint moment. Therefore, in EMG-driven models where both agonist and antagonist muscles are included, it is reasonable to assume the linear relation between the force and EMG activity of each muscle.

Model I in this study represents the linear relationship between tensile muscle force and the EMG activity of the muscle as it was used by previous investigators (Granata, 1993; Marras and Granata, 1995; McGill, 1991). In addition to the assumption of linearity of the EMG-force relationship, this model assumes that the linear relationship is explained by the linear line with zero intercept. Compared to other models, Model I resulted in significantly ($p < 0.01$) higher sagittal error and lower lateral shear force.

During this study, the lateral axis of the B200 Isostation was unlocked and therefore the subject was not able to exert moment along this axis. However, the EMG-driven model predicted moment along this axis. With the lateral axis unlocked, the values reported in Table 3.2 is the difference between the predicted moment from the EMG-driven model and the actual measured moment (zero Nm). A similar observation was reported by Hughes (1991) in a study of pure sagittal and lateral exertion. In Hughes' study, an EMG-driven model predicted sagittal moment during pure lateral exertion and lateral moment during pure sagittal exertion. This suggests a problem with these models insofar as any error in geometrical parameters or normalized EMG activity of a bilateral muscle might cause error in the predicted moment of the secondary axis.

The assumption of zero intercept in the EMG-driven models explains the specific condition of the EMG-force relationship. No justification is available in the literature for the assumption made with respect to the zero intercept in the EMG-driven models. Theoretically, it can be argued that the intercept cannot be negative. The negative intercept indicates that negative force can be generated when EMG activity is equal to zero. Therefore, the intercept can only be greater than or equal to zero. In the case of normalized EMG, it is expected that the normalization technique provides the zero intercept or reduces the intercept value close to zero. However, since the EMG is being normalized to the rest value, it is assumed that rest is equal to zero. To generate force, the muscle requires some level of EMG activity to overcome the slack part of the muscle. The level of EMG activity that is required to overcome the slack part of the musculoskeletal system is justification for the positive value of the intercept.

Model II is an expansion of Model I, where the intercept of nonzero can be assumed in the formulation of force. Compared to Models I and III, Model II suggests significantly higher compression and AP shear force. Figure 3.1 shows the sagittal moment as predicted by Models I and II and the measured moment. Model I resulted in a 35 Nm baseline shift from measured moment. In Model II, where the intercept was included, the baseline shift was reduced. This suggests that the small intercept value has a significant effect on the estimated moments. An inspection of Equation (6) indicates that the lever arm of a muscle is multiplied by the intercept as well as by the cross-sectional area and EMG value.

Both Model I and Model II assume a similar EMG-force relationship for all trunk muscles as was suggested by previous studies. Model III and Equation (4) represent a more general situation where gain and intercept can be predicted for different groups of muscles or for each muscle individually. The main reason for this consideration is twofold.

First, review of literature by Perry and Bekey (1981), suggested that many factors can effect EMG-force relationship, i.e., EMG force recording techniques, electrode type and placement, and testing methodology. Previous EMG-driven models account for the effect of muscle length and velocity in the model. The effect of other factors such as electrode type and fatigue condition can be controlled by the testing methodology. Woods and Bigland-Ritchie (1983) attempted to determine whether the EMG-force relationship is more sensitive to differences in the surface electrode (unipolar and bipolar) and recording techniques, as suggested by Moritani and DeVries (1978), or the physiological and/or anatomical differences of the motor unit organization and the muscles used in their study. Under experimentally uniform testing

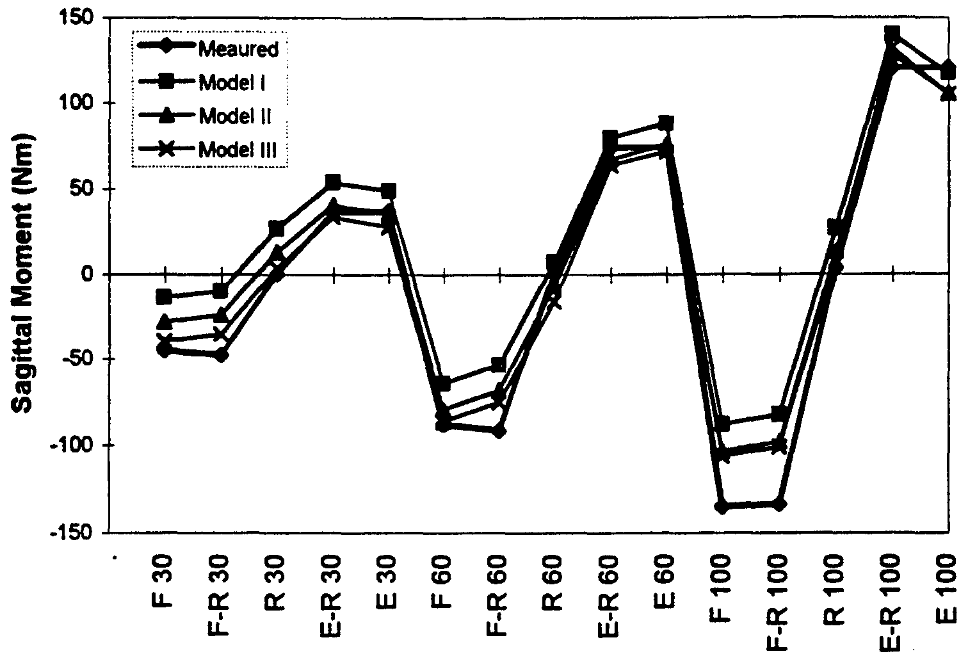

Figure 3.1. The measured sagittal moment and results of predicted sagittal moments based on Models I, II And III for pure and combined testing conditions of one subject. The horizontal axis shows flexion (F), extension (E), and right transverse (R) exertions and their combination for 30%, 60%, and 100 % MVE Exertion. The vertical axis shows the sagittal moment (+ extension, - is flexion).

conditions, muscles of different size, function, and fiber compositions were examined and then compared to conditions where the electrode type or placements had been altered. Woods and Bigland-Ritchie (1983) suggest that during isometric MVE, the EMG-force relationship depends more on the physiological characteristics of a particular muscle than the methodology employed by the investigator. They observed a linear EMG-force relationship for muscles with 'near uniform' fiber composition. The nonlinear relation was obtained from muscles of mixed fiber composition despite the electrode placement and type of electrode. These results are important for this study as the trunk muscles have mixed fiber composition (Xiong et al., 1989; Johnson et al.; 1973).

Generally, the distribution of fiber type varies between the muscles of the same individual depending on the function of a muscle, muscle exercise and training, age, and gender. Based on functional, physiological, and anatomical variation among trunk muscles, it is expected to find a difference rather than a similarity in the EMG-force relationship among these muscles.

Second, EMG-driven models are sensitive to the calibration data set. The quality and method used in the data set will directly affect the predicted gain and results of the EMG-driven model. Generally, an isometric testing condition is being used to calibrate the EMG-driven models. Guizk et al. (1996) used flexion, extension, and right and left lateral trunk exertion to calculate the EMG-force relation of different sets of muscles. The estimated gain for the trunk flexor muscles were determined by dividing the measured moments by the sum of the moments estimated for the flexor muscles, i.e., the right and left RA, IOB, and EOB muscles. A similar technique was used for extension gain where the extension muscles were assumed to include the right and left LAT, ES, quadratus lumborum, and psoas major muscles. For the right and left lateral trunk flexion, all these muscles are assumed to be involved. They observed significantly different gain values for the muscles during an exertion task. A limitation of their study is the elimination of the antagonistic muscle in the calculation of gain value.

In Model III, two groups of muscles were used: trunk extensor muscle (LAT and ES) and trunk flexor muscles (IOB, EOB, and RA). The method of grouping muscles in

Chapter III

EMG-Driven Model of Trunk

3.1. Review of the Literature

Epidemiological data suggests a relation between exposure to heavy load and the incidence of low back disorder. However, the exact nature of this relationship is not known. Several methods have been suggested by biomechanicians and ergonomists to identify the risk of low back injury based on information gathered from tasks and individual methods of performance. The identification of a hazardous task can be based on a simple ergonomic checklist (Halpern, 1992, Mital, 1996) or on advanced three-dimensional biomechanical models. The criteria for the selection of methods and tools used for identifying hazardous tasks is mainly dependent on the accuracy and reliability required to estimate the risk of injury.

Ergonomic checklists (Mital, 1996) inspect the existence of risk factors by applying general biomechanical and ergonomical concepts. The content and underlying theoretical framework is usually derived from expert opinion. Therefore, a ergonomic checklist can provide a simple but less accurate estimation of job or task effort and the risk of injury.

Biomechanical models provide a more accurate estimation of load exposure and thereby risk of injury by estimating the net internal forces. Various static (Chaffin, 1969; Chaffin and Erig, 1991; Hans et al., 1991; Ladin et al., 1991; McGill, 1991; Schultz et al., 1987) and dynamic (Freivalds et al., 1984; Marras and Granata, 1995; Marras and Sommerich, 1991) models are proposed to predict the net force on the spine. The rationale for the selection of an adequate biomechanical model depends on the complexity of the task, the reliability and validity required for output information, the simplicity and reliability of gathering input information, the sensitivity of the model to input information, and the acceptability of the assumptions made by the model.

The two-dimensional static models are based on equivalent muscles so that the unknown muscle forces will be equal to the number of equations available that satisfy the equilibrium conditions. These models calculate the compressive and shear forces on the spine, mainly at the L3-L4 level (Chaffin, 1969; Chaffin and Erig; 1991). These models do not consider individual muscle forces in the prediction of the load on the spinal structure. Therefore, they are not adequate for estimating the internal load of the trunk during asymmetrical exertion where the internal force increases due to high coactivation of muscles while the external load is fixed. These models offer a less complex calculation of the estimating load distribution on the spine structure. However, the reliability of estimated force decreases as the coactivation of trunk muscles increases during task performance.

The expansion of the two-dimensional models leads to indeterminate models and redundant systems. For these models, a solution is obtained by defining system constraints and formulating an objective function that can be utilized by the optimization technique. For formulation of the system equation for the trunk, the external moment of the system needs to be balanced by the internal moments. The internal moments are based on the description of muscle forces and their moment arms. For the optimization model, the constraints of the system are based on the maximum muscle exertion and the biomechanical threshold of the soft tissues. The objective function should be derived based on the biomechanical principles and neuromuscular criteria of the human performance. However, these criteria are poorly known. Most of the proposed objective functions are based on minimizing the compression and shear forces, muscle stresses, or by some combination of these parameters (Brand et al., 1986; Crowninshield and Brand, 1981; Genaidy and Houshyar, 1989; Han et al., 1991; Parnianpour, 1991; Schultz et al., 1983). Depending on the objective function, the estimated spinal forces can vary significantly. The reliability of these models depends solely on the reliability of the objective function and the anatomical information used in the model.

An EMG-driven model estimates the compressive and shear forces on the spine by calculating the resultant forces of all the trunk muscles. The individual muscle forces are estimated from EMG activity. These models are found to be less adequate for predicting axial rotation (Marras and Granata, 1995; McGill, 1991), but they produce reasonable results for tasks within the sagittal plane (McGill and Norman, 1987).

The reliability of estimated force and moments of the trunk by EMG-driven models is dependent upon the reliability and accuracy of the anatomical parameters of the trunk muscles (cross-sectional area, lever arm, and line of action), EMG signals, and the assumptions made with respect to the EMG-force relationship.

A majority of the previous models assumed a linear EMG-force relationship in the formulation of these models. Both evidence of linear (Lippold, 1952; Woods and Bigland-Ritchie, 1983) and nonlinear (Komi and Buskirk, 1970, Zuniga and Simons, 1969; Lawrence and De Luca, 1983) EMG-relationships exist in the literature. Furthermore, they assume that the linear relationship is similar for all trunk muscles. The purpose of this study is to: a) present a general EMG-driven model where the individual EMG-force relation can be predicted for each muscle, b) compare the results of the EMG-driven models for three selected assumptions with respect to the EMG-force relationship, and c) test for the hypothesis that the results of EMG-driven models will be affected significantly by the assumptions made with respect to the EMG-force relationship of the trunk muscles.

3.2. Method

Four male and two female subjects between the ages of 20-40 years with no history of low back pain for the last six months volunteered to participate in this study. All subjects were screened for neurological or physiological conditions that would make them unsuitable for this study. Screening was done using a questionnaire and interview.

3.2.1. Instruments

A multichannel data acquisition system was developed in-house to simultaneously collect 16 channel EMG signals and 16 channel mechanical data from B200 Isostation (Isotechnologies Inc., Hillsborough, NC). The Labview software (National Instrument, Austin, TX) was used to collect data from data acquisition system via A/D board (ATMIO-64F, National Instrument, TX) at a sampling rate of 1000 Hz. The software was programmed to provide real-time visual feedback regarding the trunk torques measured by the B200 Isostation. More details and technical information regarding the instruments and experimental setup are provided in Chapters II and IV.

3.2.2. EMG Electrodes

Both surface and wire electrodes were used to detect myoelectric activity of the trunk muscles. The surface electrodes were bipolar, pre-gelled electrodes (Classic Medical, Muskego, WI). The bipolar wire electrodes were made in-house based on methods described by Basmajian and De Luca (1985). For the best mechanical strength and stiffness, a wire with 10% Iridium- 90% platinum alloy (Medwire, Mount Vernon, NY) was used. The wire was Teflon coated with a full coated diameter of 0.0014 inches. To make the wire electrode, the 25 cm Teflon coated wire was looped and passed through cannula of 26 G X 3.5" hypodermic spinal tape needle (Baxter Health Care Co., Edison, NJ). Then the loop was cut into two pieces of wire. About 2 mm of each wire at the distal end of the needle was exposed to flame in order to burn the Teflon isolation. The electrodes were sterilized in the Hospital for Joint Diseases before use.

3.2.3. Procedure

All subjects were asked to fill out the activity questionnaire (Baecke et al., 1982), demographic and medical questionnaire (see Appendices A and B), and sign the consent form (see Appendix E). The demographic and medical questionnaires were reviewed to ensure that the subject satisfied the inclusion criteria. The subject was asked to wear comfortable sport wear and shoes during testing.

Step 1. Electrode placement

Both wire and surface electrodes were used to collect EMG signals from five bilateral trunk muscles. Wire electrode was used for the right and left internal oblique (IOB) muscles and surface electrode was used for external oblique (EOB), erector spinae (ES), rectus abdominis (RA), and latissimus dorsi (LAT) left and right muscles. In Chapter II, the details of muscle landmarks used for surface electrode placements are reported. To insure the safety and accuracy of wire electrode placement, the wire electrodes were placed using needle biopsy procedures under CT-scan by a board certified radiologist. Upon completion of the questionnaire, the subject was scheduled for a CT-scan (CT 9800 HiLight) at the Radiology Department, Hospital for Joint Diseases, New York University Medical Center.

Each subject was placed in a supine position on the CT-scan table. Before and during the procedure, the subject was instructed to relax and maintain his/her position. The L3/L4 level was verified from the sagittal plane image of the trunk. Two 10 mm slice thickness images were obtained at the L3/L4 level in order to optimally localize the internal oblique muscles in the transverse plane. This location was marked on the subject's skin with ink, and radiopaque beads were placed at 1 cm intervals bilaterally in this transverse plane above the expected location of the internal oblique muscles. A CT image showing the marker beads on the skin allowed a selection of the correct sagittal plane and therefore the best point on the skin surface to advance the electrodes. At this level, markers were placed on the skin. A CT-scan was then taken at the L3/L4 level in the transverse plane. Based on the marker placement and location of the internal oblique muscle, the site of insertion of the wire electrode was selected.

Following sterile preparation, with CT scan guidance, the electrodes were slowly placed into the internal oblique muscle. No local anesthetic was necessary. Because the electrodes could not be pulled back and re-advanced, it was essential to slowly advance each needle with repeated CT imaging to check the position. In most cases, the electrode reached the mid-third of cross-sectional area of IOB muscle within a three-step advancement of the electrode. Once the electrode reached the mid-third of cross-sectional area of IOB muscle, the spinal needle was removed and the wires were inspected and taped to the body.

Upon the completion of the wire electrode placement, the subject was then brought to the Muscle Recruitment Laboratory in the hospital. The site of surface electrode placement for four bilateral muscles, EOB, RA, ES at L3/L4 and at L1-T12 level for LAT, was prepared, based on anatomical considerations, palpation of muscles during contraction, and the recommendations of previous studies (Gabriel, 1992; Ross et al., 1993; Schmitz, 1992; Tan et al., 1993). Further details with regard to the electrode placement sites are provided in Chapter II.

Step 2. Maximum Voluntary Exertion (MVE)

The participant was strapped to the B200 Isostation in an upright standing posture according to the method described in the B200 User's Manual (Isotechnologies, Inc.,

1988) and the procedure described in previous studies (Gabriel, 1992; Ross et al., 1993; Schmitz, 1992; Tan et al., 1993). The subject performed two trials of MVE of the trunk in an upright standing posture in four random directions: flexion, extension, and right and left axial rotation. The secondary axes were unlocked during all the trials. The subject was asked to reach to his/her MVE and hold it for 3 seconds. Two minutes of rest separated each test to avoid fatigue.

Step 3. Pure and combined exertion

Based on the MVE values, the computer randomly selected a test from combinations of 0, 30, 60, and 100% of MVE for flexion, extension, and rotation of the trunk. To simulate pure and combined exertions, visual feedback was given to the subjects. A monitor visually displayed trunk torque output and target torque in the plane of sagittal and axial torque. The lateral axis was unlocked, hence provided no resistance along this axis. The computer provided an audio warning if a subject exceeded ±5 degree of lateral bending. A target torque was set at 0, 30, 60, 100% of MVE exertion along one or combination of two axes of the trunk. Each subject was instructed to match their output torque to the target torque for at least three seconds. To reduce the number of testing conditions and the risk of fatigue, only rotation to the right side is considered in this study (see Figure 2.4 for test condition used in this study). No load condition, zero condition along each axis, was repeated more than three times. However, these tests were part of the randomization process. The zero condition served as a means to calculate the baseline EMG for the normalization of the data.

Before each test, the feedback screen was displayed on the terminal and the subject was asked to practice reaching the target area. Based on the subject's performance, verbal feedback was provided. The practice trial was continued until the subject felt comfortable with the task and indicated that he/she was ready to start the test.

3.2.4. Data Analysis

3.2.4.1. Data reduction

Based on Equation (1), three seconds of each trial was selected in which the subject performed closest to the target torque. The close distance was defined by the low value of the control index as it was calculated by the following formula:

$$\text{Control Index} = \frac{\sqrt{\left(Sag_{Torque} - Sag_{Tag}\right)^2 + \left(Tran_{Torque} - Tran_{Tag}\right)^2}}{\sqrt{\left(Sag_{Tag}\right)^2 + \left(Tran_{Tag}\right)^2}} \tag{1}$$

Where :

Sag_{Torque} and $Tran_{Torque}$ = The generated sagittal (Sag) and transverse (Tran) torque

Sag_{Tag} and $Tran_{Tag}$ = The sagittal (Sag) and transverse (Tran) target torque

For all ten trunk muscles, the mean of the normalized Root Mean Square of the EMG signal (NRMS-EMG) was calculated for a selected 3 second window. Based on previous recommendations (Redfern, 1992), the time constant of 50 ms was used for calculation of the RMS. As suggested by Lavender et al. (1992) and Seroussi and Pope (1987), the EMG signal of each muscle was normalized to the maximum and the rest value of the muscle. In this study, the highest mean NRMS-EMG values among all the trials were selected for the normalization procedure.

3.2.4.2. CT-SCAN

The CT-scan images were down loaded from the CT-scan unit to optical disks. At the Radiology Department of the School of Medicine, New York University, the images were converted to 512x512 binary image files. Sigma Scan Pro image processing software (Jandel Scientific, San Rafael, CA) was used to process the CT-scan images. From the CT-scan images, the width and depth of trunk, cross-sectional area, and distance of centroid of muscle from the center of the disk were measured. The sagittal view of the spine was inspected to insure that the angle of L3/L4 vertebra with respect to the horizontal line was about 90°, and that the values measured from these images did not need any correction, as suggested by McGill et al. (1993).

3.2.4.3. Calculation of Muscle Forces

The EMG-driven model was used to predict trunk moments and spinal forces. The EMG-driven model is based on the assumption that the tensile force generated by a muscle is related to the normalized EMG and cross-sectional area of a muscle. There are controversial reports with respect to the characterization of this relationship as linear (Lippold, 1952; Woods and Bigland-Ritchie, 1983) and nonlinear (Komi and Buskirk, 1970, Lawrence and De Luca, 1983; Zuniga and Simons, 1969). In this study as well as others (Granata, 1993), a linear relation between tensile force and EMG has been employed, based on several factors: 1) the nonlinear relationship for the mid-range of the force can be approximated as linear (Granata, 1993; Solomonow et al., 1986); 2) when the coactivation of muscles was accounted for in a model, based on Hof and Van Den Berg (1977), the EMG-force relationship was linear; and 3) for simplicity of formulation, the relationship can be assumed to linear and may be expanded to be nonlinear in the future.

If the relation between the myoelectrical activity of muscle and tensile force is assumed to be linear, then the tensile force (F_i) of muscle i is:

$$F_i = G_i \times (A_i \times NEMG_i) + b_i \tag{2}$$

where:

G_i = Gain of ith muscle
$NEMG_i$ = Mean of normalized RMG-EMG of ith muscle
A_i = Cross-sectional area of ith muscle
b_i = intercept of line

Since the muscle force cannot be measured directly, G_i can be calculated from moment equations, where the measured moment (M) for the j axis is:

$$M_j = \sum_{i=1}^{n} \left(G_i \times (A_i \times NEMG_i) + b_i\right) \times L_{ij} \tag{3}$$

$$M_j = \sum_{i=1}^{n} \left(G_i \times A_i \times NEMG_i \times L_{ij}\right) + \left(b_i \times L_{ij}\right) \tag{4}$$

where L_{ij} is the lever arm of muscle i in j plane, and n is the number of muscles, which in this study is equal to ten.

In this study, three models have been formulated based on the assumptions that can be made with respect to G_i and b_i.

Model I: Based on previous studies, two assumptions can be made:

a) The gain of all muscles is equal to a constant G value:

$$G = G_1 = G_2 = \ldots = G_{10}$$

b) The intercept of all muscles is equal to zero:

$$b_1 = b_2 = \ldots = b_{10} = 0$$

Based on these two assumptions, Equation (4) can be simplified as:

$$M_j = G \times \left(\sum_{i=1}^{10} A_i \times NEMG_i \times L_{ij} \right) \qquad (5)$$

Model II: Similar to Model I, this model assumes that the gain of all trunk muscles is equal to a constant G value. Furthermore, it assumes that b_i of all the muscles is equal to each other:

$$b = b_1 = b_2 = \ldots = b_{10}$$

Based on this assumption, Equation (4) can be simplified as:

$$M_j = \sum_{i=1}^{10} \left(G \times A_i \times NEMG_i \times L_{ij} \right) + \left(b \times L_{ij} \right) \qquad (6)$$

Model III: Specific conditions can be assumed; that the trunk muscles can be grouped together based on their functional (i.e., flexors vs. extensors) or physiological (i.e., fiber types) properties. For instance, in Model III, we assume that:

$$M_j = \sum_{p=1}^{m} M_{jp} \qquad (7)$$

where :

$$M_{jp} = \sum_{i=1}^{n_p} \left(G_p \times A_i \times NEMG_{i_p} \times L_{ij_p}\right) + \left(b_p \times L_{ij_p}\right) \tag{8}$$

m is the number of grouped muscles and it can take any value from one to ten. Equations (4) and (7) present cases in which m is assumed to be one and ten respectively. In this model, we assume two groups of muscles: 1) extensor muscles that include the right and left LAT and ES muscles , and 2) flexor muscles that include the right and left RA and IOB and EOB muscles. For calculation of muscle force in these three models, the line of action of the muscle was used as reported by Nussbaum et al. (1995).

3.2.4.4. Ridge Regression

A detailed explanation of the ridge regression method and its derivation is presented by Neter et al. (1983). The following is brief summary of this method. For ordinary least squares,

$$y = bx \tag{9}$$

Neter et al., (1983) showed that the regression coefficient can be calculated by:

$$b = \left(r_{xx}\right)^{-1} r_{yx}$$

Where r_{xx} is a correlation matrix of all the independent variables, and r_{yx} is the vector containing the coefficients of simple correlation between the dependent variable and each of the independent variables.

The ridge regression estimates are obtained by (Neter et al., 1983):

$$b^r = (r_{xx} + CI)^{-1} r_{yx} \tag{10}$$

where C is a bias constant ($C \geq 0$), b^r is the vector of standardized ridge regression coefficient , and I is identity matrix.

The value of C indicates the amount of bias in the estimators. The value of C can range from zero to one. When C is equal to zero, Equation (10) is the ordinary least square regression. Neter et al. (1983) indicate that as C increases from zero to one, the total mean squared error of the ridge regression estimator b^r increases, while the variance component decreases.

Generally, the ridge trace is used to estimate the C value. The ridge trace is a plot of the b^r value (Equation (10)), when C takes a value between the range of zero to one. The ridge trace usually has a large fluctuation for the slight change of the C value from zero, and then tends to be stabilized for higher C values. By the visual inspection of the trace, one should select the smallest value of C where all the traces are relatively stable. In this study, to reduce the risk of error due to visual inspection of the ridge trace, we selected the value of C=0.01 for all the models. This value was selected since there is no criteria available for proper selection of C value any pre-select method might produce systematic error.

To solve Equation (9) by means of ridge regression, the data was arranged for each axis as follows:

$$M_j = \begin{bmatrix} a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \end{bmatrix} \begin{bmatrix} G_1 \\ \vdots \\ G_n \\ b_1 \\ \vdots \\ b_n \end{bmatrix} \qquad (11)$$

where:

$a_{mn} = A_{mn} \times NEMG_{mn} \times L_{mn}$ for n^{th} muscle in trial m.
ℓ_{mn} = Lever arm of the n^{th} muscle in trial m
G_n = the gain of n^{th} muscle
b_n = the intercept of n^{th} muscle

To calculated the gain and intercept based on three axes, the Equation (11) should be modified to include the data of all three axis in ridge regression model. Equation (12) shows the expansion of Equation (11) where the data of all three axis were appended to a large matrix.

$$\begin{bmatrix} M_{sagittal} \\ M_{Transverse} \\ M_{Lateral} \end{bmatrix} = \begin{bmatrix} a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \\ a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \\ a_{11} & \cdots & a_{1n} & \ell_{11} & \cdots & \ell_{1n} \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ \vdots & \cdots & \vdots & \vdots & \cdots & \vdots \\ a_{m1} & \cdots & a_{mn} & \ell_{m1} & \cdots & \ell_{mn} \end{bmatrix} \begin{bmatrix} G_1 \\ \vdots \\ G_n \\ b_1 \\ \vdots \\ b_n \end{bmatrix} \qquad (12)$$

3.2.5. Statistical Analysis

For each subject, the mean of absolute error (the difference between the predicted and the actual moment produced) of trunk moments, compression, and shear forces were calculated. Descriptive statistics (the mean and standard deviation) were computed for demographic data, absolute error of moments, compression, and shear forces at the level of L3/L4 of the spine.

The statistical procedure MANOVA of repeated measure design (SAS Institute Inc., 1985) was used to determine the effect of the method of calculation (models) used on the results of the EMG-driven model of trunk. If the overall multivariate analysis was significant, a univariate analysis (ANOVA) was performed. The Tukey test was performed on the levels of the model if the effect of the univariate test was statistically significant. In this study, statistical significant level was set at $P<0.05$.

3.3. Results

The CT-scan and EMG data of 6 subjects (4 males and 2 females) was used in the analysis. Mean age, weight, and height were 27.3 (±5.4) years, 68.0 (±6.9) Kg, and 169 (±8.8) cm, respectively. The scores for index of work, sport, and leisure were 2.40 (±0.66), 2.95 (± 1.68), and 3.25 (±0.57) respectively. Table 3.1 summarizes the mean and standard deviation of the muscle parameters of the six subjects based on values measured from the CT-scan each individual subject. Additional data from ten subjects were collected but not included in this data analysis due to different technical reasons related to either CT-scan data (i.e., low resolution for separating the muscle or the loss of calibration information of the image), or missing EMG signal data (i.e., resulting from the loss of the wire or surface electrode signal), (Chapter II for detailed information).

For each subject, the mean of absolute error across all the trials was calculated. Table 3.2 shows the mean of absolute error for each plane, and the mean of spinal forces for all subjects. The sagittal moment error was 26.0 (± 4.7) Nm for Model I, and was reduced to 22.3 (± 9.2) Nm and 17.5 (± 5.7) Nm for Models II and III respectively. Similar level of errors were observed for the results of the three models with respect error in predication of transverse and lateral moments.

With respect to spinal forces, the three models differed in their ability to predict compression forces and lateral shear forces. The results of Model II predicted the highest lateral shear and anterior-posterior (AP) shear and compression forces compared to the other two models. Model III resulted in the lowest compression force with a mean of 1614.02 (± 796.5) N.

The result of the multivariate analysis of variance indicated a significant ($p<0.001$) effect of the models on the results of the EMG-driven model. The repeated ANOVA was performed to investigate the effect of the model on the mean of the predicted moments and spinal forces. Table 3.2 shows the summary of the results of the ANOVA procedure.

Table 3.3 presents the results of the post-hoc Tukey test for the main effect of the model on the mean of sagittal error, lateral shear and compression forces. Model I had significantly higher ($p <0.0049$) sagittal error compared to Model III. With respect to

Table 3.1. Mean and standard deviation of the cross-sectional area and the lever arm (the distance of the centroid of the muscle to the center of disc) for ten selected trunk muscles at the L3/L4 level of the spine.

	Lateral Lever Arm (mm)	Anterior-posterior Lever Arm (mm)	Cross-Sectional Area (mm^2)
Left ES	37.32 (3.81)	59.54 (4.59)	2506.79 (443.61)
Right ES	35.49 (7.94)	59.17 (5.63)	2540.37 (400.20)
Left RA	41.91 (10.12)	74.33 (13.02)	855.81 (201.36)
Right RA	44.93 (3.0)	75.02 (15.81)	884.83 (168.17)
Left IOB	108.42 (11.13)	17.81 (8.06)	1458.91 (406.86)
Right IOB	110.78 (11.00)	19.32 (11.32)	1317.51 (335.85)
Left EOB	121.2 (10.83)	15.2 (10.06)	1335.25 (357.79)
Right EOB	122.29 (10.15)	16.19 (12.02)	1320.09 (239.49)
Left LAT	102.56 (11.61)	39.93 (11.83)	193.62 (54.46)
Right LAT	102.53 (10.01)	39.62 (14.99)	215.63 (52.97)

lateral shear force, Model I predicted significantly (p <0.0052) lower lateral shear force. Significantly (p<0.02) higher compression force was predicted by Model II in comparison to the other two models.

Table 3.2. Mean and standard deviation for results of EMG-driven models and univariate analysis of variance (ANOVA) for the mean of the absolute predicted moments and compression and shear forces.

	Model I	Model II	Model III	F and P Value
Sagittal Error (Nm)	26.0 (4.7)	22.3 (9.2)	17.5 (5.7)	6.51 (0.015)
Transverse Error (Nm)	19.9 (7.5)	18.6 (8.8)	19.1 (8.1)	NS
Lateral Error (Nm)	14.1 (6.2)	15.6 (5.0)	13.9 (4.1)	NS
Lateral Shear (N)	66.7 (25.2)	72.0 (25.5)	72.4 (25.9)	7.90 0.008)
AP Shear (N)	448.7 (148.2)	495.7 (181.4)	424.8 (96.4)	NS
Compression (N)	1631.4 (718.0)	3961.3 (2597.9)	1614.02 (796.5)	4.79 0.035

Table 3.3. The P values of the Tukey multiple comparison test for main effect of model on the means of EMG-driven model of the trunk.

	Model I Vs II	Model I Vs III	Model II Vs III
Sagittal Error	-	1>3 0.0049	-
Lateral Shear	1 < 2 0.0078	1 > 3 0.0052	-
Compression	1 < 2 0.0234	-	2 > 3 0.0226

Table 3.4 Predicted gain (N/m^2) and intercept (Nm) based on three EMG-driven models for each subject (ID) and the mean and standard deviation (STD) for six subjects.

	Model 1	Model II		Model III			
ID	Gain	Gain	Intercept	Flexor Gain	Extensor Gain	Flexor Intercept	Extensor Intercept
1	0.55	0.55	0.1116	0.59	0.53	-0.0826	-0.2322
2	0.28	0.36	0.8658	0.33	0.40	0.2259	0.0453
3	0.44	0.51	-0.4224	0.58	0.45	-0.1014	-0.1428
4	0.97	1.04	-0.8132	1.09	1.05	-0.5357	-0.6668
5	0.57	0.60	-0.3334	0.69	0.56	-0.0788	-0.0823
6	0.43	0.43	0.2929	0.63	0.41	-0.0082	-0.0494
Mean	0.54	0.58	-0.04987	0.65	0.57	-0.09683	-0.1880
STD	0.23	0.24	0.5973	0.25	0.24	0.2470	0.2522

Table 3.4 demonstrates the estimated gain and intercept of the EMG-force relationship as predicted by the ridge regression model. The lowest gain across all three models was 0.28 N/m^2 and the highest was 1.09 N/m^2. The mean of predicted gain ranged from 0.54 to 0.65 N/m^2. Subject 4 had the highest gain value across all three models. The intercept estimated for Models II and III ranged from -0.813 Nm to 0.865 Nm across all subjects.

3.4. Discussion

The statistical analysis suggests that the results of the EMG-driven models are significantly affected by the methods of muscle force calculation. The three mathematical formulations of force in this study stem from the possible theoretical assumptions made with respect to the EMG-force relationship.

Many studies have been performed to determine the relationship between surface EMG and the force generated by the muscle during various type of contraction. The

EMG-force relationship has been shown to be linear (Lippold, 1952; Woods and Bigland-Ritchie, 1983) and nonlinear (Komi and Buskirk, 1970; Lawrence and De Luca, 1983; Zuniga and Simons, 1969). The discrepancy with respect to the reported relation is mainly due to a variation of investigated muscles and the method used by the investigator (Perry and Bekey, 1981).

In addition to methodological differences in collecting force and EMG data , Hof and Van Den Berg (1977) demonstrated that the EMG of a muscle is linearly related to muscle force, but coactivation muscles generates a nonlinear relation between the EMG and the joint moment. Therefore, in EMG-driven models where both agonist and antagonist muscles are included, it is reasonable to assume the linear relation between the force and EMG activity of each muscle.

Model I in this study represents the linear relationship between tensile muscle force and the EMG activity of the muscle as it was used by previous investigators (Granata, 1993; Marras and Granata, 1995; McGill, 1991). In addition to the assumption of linearity of the EMG-force relationship, this model assumes that the linear relationship is explained by the linear line with zero intercept. Compared to other models, Model I resulted in significantly ($p < 0.01$) higher sagittal error and lower lateral shear force.

During this study, the lateral axis of the B200 Isostation was unlocked and therefore the subject was not able to exert moment along this axis. However, the EMG-driven model predicted moment along this axis. With the lateral axis unlocked, the values reported in Table 3.2 is the difference between the predicted moment from the EMG-driven model and the actual measured moment (zero Nm). A similar observation was reported by Hughes (1991) in a study of pure sagittal and lateral exertion. In Hughes' study, an EMG-driven model predicted sagittal moment during pure lateral exertion and lateral moment during pure sagittal exertion. This suggests a problem with these models insofar as any error in geometrical parameters or normalized EMG activity of a bilateral muscle might cause error in the predicted moment of the secondary axis.

The assumption of zero intercept in the EMG-driven models explains the specific condition of the EMG-force relationship. No justification is available in the literature for the assumption made with respect to the zero intercept in the EMG-driven models. Theoretically, it can be argued that the intercept cannot be negative. The negative intercept indicates that negative force can be generated when EMG activity is equal to zero. Therefore, the intercept can only be greater than or equal to zero. In the case of normalized EMG, it is expected that the normalization technique provides the zero intercept or reduces the intercept value close to zero. However, since the EMG is being normalized to the rest value, it is assumed that rest is equal to zero. To generate force, the muscle requires some level of EMG activity to overcome the slack part of the muscle. The level of EMG activity that is required to overcome the slack part of the musculoskeletal system is justification for the positive value of the intercept.

Model II is an expansion of Model I, where the intercept of nonzero can be assumed in the formulation of force. Compared to Models I and III, Model II suggests significantly higher compression and AP shear force. Figure 3.1 shows the sagittal moment as predicted by Models I and II and the measured moment. Model I resulted in a 35 Nm baseline shift from measured moment. In Model II, where the intercept was included, the baseline shift was reduced. This suggests that the small intercept value has a significant effect on the estimated moments. An inspection of Equation (6) indicates that the lever arm of a muscle is multiplied by the intercept as well as by the cross-sectional area and EMG value.

Both Model I and Model II assume a similar EMG-force relationship for all trunk muscles as was suggested by previous studies. Model III and Equation (4) represent a more general situation where gain and intercept can be predicted for different groups of muscles or for each muscle individually. The main reason for this consideration is twofold.

First, review of literature by Perry and Bekey (1981), suggested that many factors can effect EMG-force relationship, i.e., EMG force recording techniques, electrode type and placement, and testing methodology. Previous EMG-driven models account for the effect of muscle length and velocity in the model. The effect of other factors such as electrode type and fatigue condition can be controlled by the testing methodology. Woods and Bigland-Ritchie (1983) attempted to determine whether the EMG-force relationship is more sensitive to differences in the surface electrode (unipolar and bipolar) and recording techniques, as suggested by Moritani and DeVries (1978), or the physiological and/or anatomical differences of the motor unit organization and the muscles used in their study. Under experimentally uniform testing

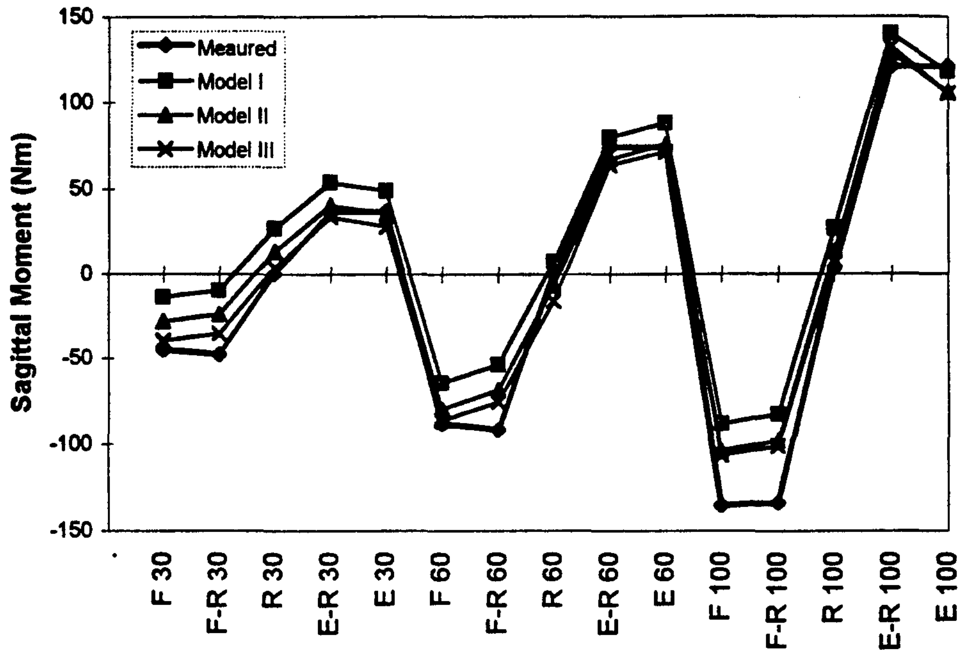

Figure 3.1. The measured sagittal moment and results of predicted sagittal moment based on Models I, II and III for pure and combined testing conditions of one subject. The horizontal axis shows flexion (F), extension (E), and right transverse (R) exertions and their combination for 30%, 60%, and 100 % MVE. The vertical axis shows the sagittal moment (+ extension, - flexion).

conditions, muscles of different size, function, and fiber compositions were examined and then compared to conditions where the electrode type or placements had been altered. Woods and Bigland-Ritchie (1983) suggest that during isometric MVE, the EMG-force relationship depends more on the physiological characteristics of a particular muscle than the methodology employed by the investigator. They observed a linear EMG-force relationship for muscles with 'near uniform' fiber composition. The nonlinear relation was obtained from muscles of mixed fiber composition despite the electrode placement and type of electrode. These results are important for this study as the trunk muscles have mixed fiber composition (Zhu et al., 1989; Johnson et al.; 1973).

Generally, the distribution of fiber type varies between the muscles of the same individual depending on the function of a muscle, muscle exercise and training, and age. The size of different fiber types varies somewhat according to sex, but not the proportion of the fiber types (Zhu et al., 1989). Johnson et. al. (1973) studied samples of skeletal muscles taken from 50 sites of 6 cadavers. The cadavers were between age of 17 to 30 years old with no previously known medical conditions. They reported wide variation between percentage of Type I and Type II fibers among the specimens. With regard to trunk muscles, the proportion of Type I and Type II fiber of ES (both at surface and deep level), LAT, and RA were significantly ($p<0.002$) different among the specimens. Therefore, considering difference among individuals, and the functional, physiological, and anatomical variation among trunk muscles, it can be expected to find more difference rather than a similarity in the EMG-force relationship among the trunk muscles.

Second, EMG-driven models are sensitive to the calibration data set. The quality and method used in the data set will directly affect the predicted gain and results of the EMG-driven model. Generally, an isometric testing condition is being used to calibrate the EMG-driven models. Guizk et al. (1996) used flexion, extension, and right and left lateral trunk exertion to calculate the EMG-force relation of different sets of muscles. The estimated gain for the trunk flexor muscles were determined by dividing the measured moments by the sum of the moments estimated for the flexor muscles, i.e., the right and left RA, IOB, and EOB muscles. A similar technique was used for extension gain where the extension muscles were assumed to include the right and left LAT, ES, quadratus lumborum, and psoas major muscles. For the right and left lateral trunk flexion, all these muscles are assumed to be involved. They observed significantly different gain values for the muscles during an exertion task. A limitation of their study is the elimination of the antagonistic muscle in the calculation of gain value.

In Model III, two groups of muscles were used: trunk extensor muscle (LAT and ES) and trunk flexor muscles (IOB, EOB, and RA). The method of grouping muscles in this model was primly done to demonstrate a possibility of logical grouping of muscles. Compared to Models I and II, Model III had the lowest sagittal and lateral mean error (Table 3.2). The mean of flexor muscle gain as predicted by the Model III was 65 (±25) N/cm^2 and 57(±24) N/cm^2 for the extensors. Similar to Guzik et al. (1996), we found a higher gain for the trunk flexor than for the trunk extensor muscles. The absolute value of the gain was different due to variations in the testing methodology and muscle grouping. Their study included the psoas major and quadratus lumborum muscles in addition to the LAT and ES muscles that were studied Model III.

For all three models, a ridge regression was used to calculate the gain. The advantage of this method is twofold: first, theoretical gain can be calculated from the equation of moments and the gain value is independent of the axis used in the calculation. However, experimentally, we know that the result of the EMG-driven model is more accurate in the prediction of moments in the sagittal plane than in the transverse and lateral planes. Therefore, numerically the gain value will be different depending upon which axis is used in the calculation of gain. No explicitly information is available regarding this issue in the literature. Two numerical solutions can be assumed for this problem: 1) calculating the gain based on each axis and averaging the three estimated gains, 2) fitting the linear least square line to the data of all three axes. The latter technique requires to append the data of all three axes, as shown in Equation (11).

Second, the best method of solving Equation (4) is the use of a regression based method. However, due to the collinearity of the EMG data, the results of the regression method become biased. The ridge regression is one of the few methods that has been proposed to solve the problem of multicollinearity in regression models. This method allows researchers to choose the regression coefficient from a range where the coefficient is unbiased but imprecise (with a large variance) to a value that is more precise but has a small bias (Neter et al., 1983). The flexibility on this range will provide the possibility of selecting the regression coefficient (EMG-force ratios) such that it falls within the physiological range of gain; i.e., 30 to 100 N/cm^2 (McGill and Norman, 1987; Reid and Costigan, 1987).

Table 3.4 shows the gain and intercept as it was calculated by three EMG-driven models. The estimated gains were within the physiological range reported in the literature. One subject had constantly high gain values across all three models. Table 3.4 indicates that the predicted intercept can be either positive or negative. Theoretically, the intercept should be positive. The negative intercept might be due to the normalization technique used in the study. In this study, we normalized the EMG value of each muscle by the highest value observed across all the trial and rest EMG. However, since the subject was standing in an upright, constrained posture within the dynamometer during the resting trial, the muscles may have had some postural EMG activity.

3.5. Conclusion

An EMG-driven model has been used to estimate spinal forces during upright isometric standing. The shortcomings of previous models have been discussed. To overcome these limitations, a new method of calculating individual muscle force has been suggested. The main advantage of this proposed method is the possibility of calculating the individual gain and intercept for the trunk muscles.

The results of this study demonstrate the sensitivity of the EMG-driven models and estimated spinal forces to the assumption made with respect to individual muscle forces. The variation in the estimation of moments and spinal forces by these models suggests that one should be careful in the interpretation of these results. The model should be evaluated, based on: 1) the accuracy of the anatomical and physiological assumptions used in the model, 2) error on predicting the moment along each axis, and 3) the gain estimated by the model as well as the compression and shear forces with in the limits suggested in the literature.

3.6. References

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Chapter IV

The Effect of Anatomical Geometry of Trunk Muscles on the Results of the EMG-driven Model

4.1 Review of the Literature

The interest in biomechanical models of human performance and advanced development of computational power for research encourage the formulation and examination of more complex and rigorous biomechanical models (Pearsall and Reid, 1994). The primary objective of these models is to calculate the internal forces generated by the musculature and the estimation of risk of muscluoskeletal injury. The validity of the biomechanical models are dependent upon the physiological assumptions and the anatomical parameters used (i.e., cross-sectional area, lever arm, and centroid of the muscle) .

Until recently, biomechanical models were based on anatomical and physiological data gathered from human cadaver studies. These studies derive information from a small number of specimens. Often the cadaver specimens used were from an elderly population or from young people who died due to medical conditions (Pearsall and Reid, 1994) or accidents. There is potential for error in utilizing these information because of the differences between healthy, living humans and the cadaver specimens used in these studies (Pearsall and Reid, 1994). Furthermore, these studies are usually based upon a relatively limited demographic sample of subjects (Pearsall and Reid, 1994). Use of cadaver data might be adequate for simple biomechanical models, but more complex models of the human body require more specific and individualized data.

Two approaches have been used to improve the quality of the anatomical parameters required by the biomechanical models. First, the use of noninvasive imaging techniques provides the opportunity to gather in vivo anatomical information from healthy individuals. Individualized data can be gathered from CT-scan and Magnetic Resonance Imaging (MRI) images with a high degree of accuracy. The cost and availability of these imaging techniques are the primary limitations for their use in research projects.

The second approach for improving the quality of anatomical parameters is the use of statistical methods for calculating parameters based on anthropomorphic data. This method implies that anatomical parameters will be measured either directly from cadavers or by available imaging techniques from subjects. The demographic and anthropometric data of the same subjects will then be recorded and the relation between these variables is established via a regression model. Depending on the strength of these relationships, this method provides relatively more individualized information while keeping the process of gathering the information both inexpensive and less complex.

Reid et al. (1987) used 27 different anthropometric variables as the base for their prediction of the moment arms and the cross-sectional areas of five trunk muscles in twenty male subjects. Reid and colleagues showed a low correlation ($p>0.10$ and $p>0.15$) among anthropometric variables and the cross-sectional areas of the muscles. Two studies, Reid and Costigan (1987) study of twenty male subjects, and Chaffin et al. (1990) study of 96 women (mean age of 49.6 (±5.9)), more specifically point to the low correlation between gross anthropometry and cross-sectional areas for the erector spinae and psoas muscles. McGill and Norman (1987) showed similar results for the psoas muscle, but not for the erector spinae muscle.

Several authors (Chaffin et al., 1990; Nemeth and Ohlsen, 1986; Reid et al., 1987) attempted to predict the moment arms of the trunk muscles based on gross anthropometric parameters. Reid et al. (1987), showed that a "stable regression equation" could be developed for five trunk muscles with the exception of the moment arm for the rectus abdominus muscle. The lack of a stable equation for this muscle indicates high variability among the subjects.

A comparison of the results of several studies (Chaffin et al., 1990) reveals that the length of the moment arm of the erector spinae and rectus abdominus muscle with respect to the anteroposterior axis are highly variable among studies. A comparison of the anteroposterior lever arm and the mean age of the subjects suggests an increase in the rectus abdominus moment arm as the mean age of the population increases. The increase is estimated to be as high as 30% and appears mainly in the 30 year age cohort. The comparison was made based on data from male subjects. Little information is available on the female population, although a comparative study of ten males and ten females by Nemeth and Ohlsen (1986) showed a significant difference between the male and the female moment arm with respect to the anteroposterior and bilateral axis of motion.

In conclusion, the review of the literature suggests that muscle parameters cannot be accurately predicted from anthropometric data. Limited data with high variability on the one hand, and the need for a better mathematical model on the other hand, are the main justifications for the high cost of measurement of these parameters through non-invasive, high-tech imaging systems such as the CT-scan or MRI.

The aims of this study are to: 1) provide individualized muscle descriptions based on CT-scan images; and 2) investigate the effect of the anatomical geometry of the trunk muscles on the results of the EMG-driven model of the trunk. Furthermore, the proposed study will discuss the role of imaging techniques for biomechanical models.

4.2 Method

Four male and two female subjects between the ages of 20-40 years with no history of low back pain for the last six months volunteered for this study. All subjects were screened for neurological or physiological conditions that would make them unsuitable for this study. Screening was done by the means of questionnaires and informal interview (see Chapter II and Appendices A and B).

4.2.1 Instruments

A multichannel data acquisition system was developed in-house to simultaneously collect 16 channel EMG signals and 16 channel mechanical data from a B200 Isostation (Isotechnologies, Inc., Hillsborough, NC). The Labview software (National Instrument, Austin, TX) was used to collect data from a data acquisition system via A/D board

(ATMIO-64F, National Instrument, TX) at a sampling rate of 1000 Hz. The software was programmed to provide real-time visual feedback regarding the trunk torque measured by the B200 Isostation.

4.2.2 EMG Electrodes

Both surface and wire electrodes were used to detect the myoelectric activity of the trunk muscles. The surface electrodes were bipolar, pre-gelled electrodes (Classic Medical, Muskego, WI). The bipolar wire electrodes were made in-house based on methods described by Basmajian and De Luca (1985). For the best mechanical strength and stiffness, wire with 10% Iridium- 90% platinum alloy (Medwire, Mount Vernon, NJ) was used. The wire was Teflon-coated with a full coated diameter of 0.0014 inches. To make the wire electrode, the 25 cm Teflon-coated wire was looped and passed through a cannula of 26 G x 3.5" hypodermic spinal tape needle (Baxter Health Care Co., Edison, NJ). Then the loop was cut into two pieces of wire. About 2 mm of each wire at the distal end of the needle was exposed to a flame in order to burn the Teflon isolation. The wires were then gas sterilized according to routine procedures at the Hospital for Joint Diseases.

4.2.3 Procedure

Subjects were asked to fill out habitual physical activity questionnaires (Baecke et al. 1982), demographic and medical forms (see Appendices A and B), and to sign the consent form (see Appendix E). The demographic and medical questionnaires were reviewed to ensure that the subject satisfied the inclusion criteria. The subject were asked to wear comfortable sport wear and shoes during testing.

Step 1. Electrode Placement

Both wire and surface electrodes were used to collect EMG signal from five bilateral trunk muscles. Wire electrode was used for the internal oblique (IOB) muscle and surface electrode was used for external oblique (EOB), erector spinae (ES), rectus abdominis (RA),and latissimus dorsi (LAT) left and right muscles. More detail information for the landmark are presented in Chapter II.

To insure the safety and accuracy of wire electrode placement, the wire electrodes were placed using needle biopsy procedure under CT-scan. Upon completion of the questionnaire, the subject was scheduled for a CT-scan (CT 9800 HiLight) at the Radiology Department of the Hospital for Joint Diseases, New York University.

Each subject was placed in a supine position on the CT table. Before and during the procedure, the subject was instructed to relax and maintain his/her position. The level L3/L4 was verified from a sagittal plane image of the trunk. Two 10 mm slice thickness images were obtained at the L3/L4 level in order to optimally localize the internal oblique muscles in the transverse plane. This location was marked on the subject's skin with ink and radiopaque beads were placed at 1 cm intervals bilaterally in this transverse plane above the expected location of the internal oblique muscles. A CT image showing the marker beads on the skin allowed a selection of the correct sagittal plane and therefore the best point on the skin surface to advance the electrodes. At this level, markers were placed on the skin. A CT-scan was then taken at the L3/L4 level in the transverse plane.

Based on the marker placement and location of the internal oblique muscle, the site of the insertion of the wire electrode was selected.

Following sterile preparation, with CT-scan guidance, the electrodes were slowly placed into the internal oblique muscle. No local anesthetic was necessary. Because the electrodes could not be pulled back and re-advanced, it was essential to slowly advance each needle with repeated CT imaging to check their position. In most cases, the electrode reached the mid-third of IOB muscle within a three steps advancement of the electrode. Once the electrode reached the mid-third of IOB, the spinal needle was removed and the wires were inspected and taped to the body. Figure 4.1 shows the location of the wire electrode in both left and right IOB muscles.

Upon the completion of wire electrode placement, the subject was then brought to the laboratory. The site of surface electrode placement for four bilateral muscles, EOB, RA, ES at the L3/L4 level and at the L1-T12 level for the LAT, was prepared based on anatomical considerations, palpation of muscles during contraction, and recommendations of previous studies (Gabriel 1992; Ross et al., 1993; Schmitz, 1992; Tan et al., 1993), see also Chapter II.

Step 2. Maximum Voluntary Exertion (MVE)

The subject was strapped to the B200 Isostation in an upright standing posture according to the method described in the B200 User's Manual (Isotechnologies, Inc., 1988) and the procedure described in previous studies (Gabriel, 1992; Ross et al., 1993;

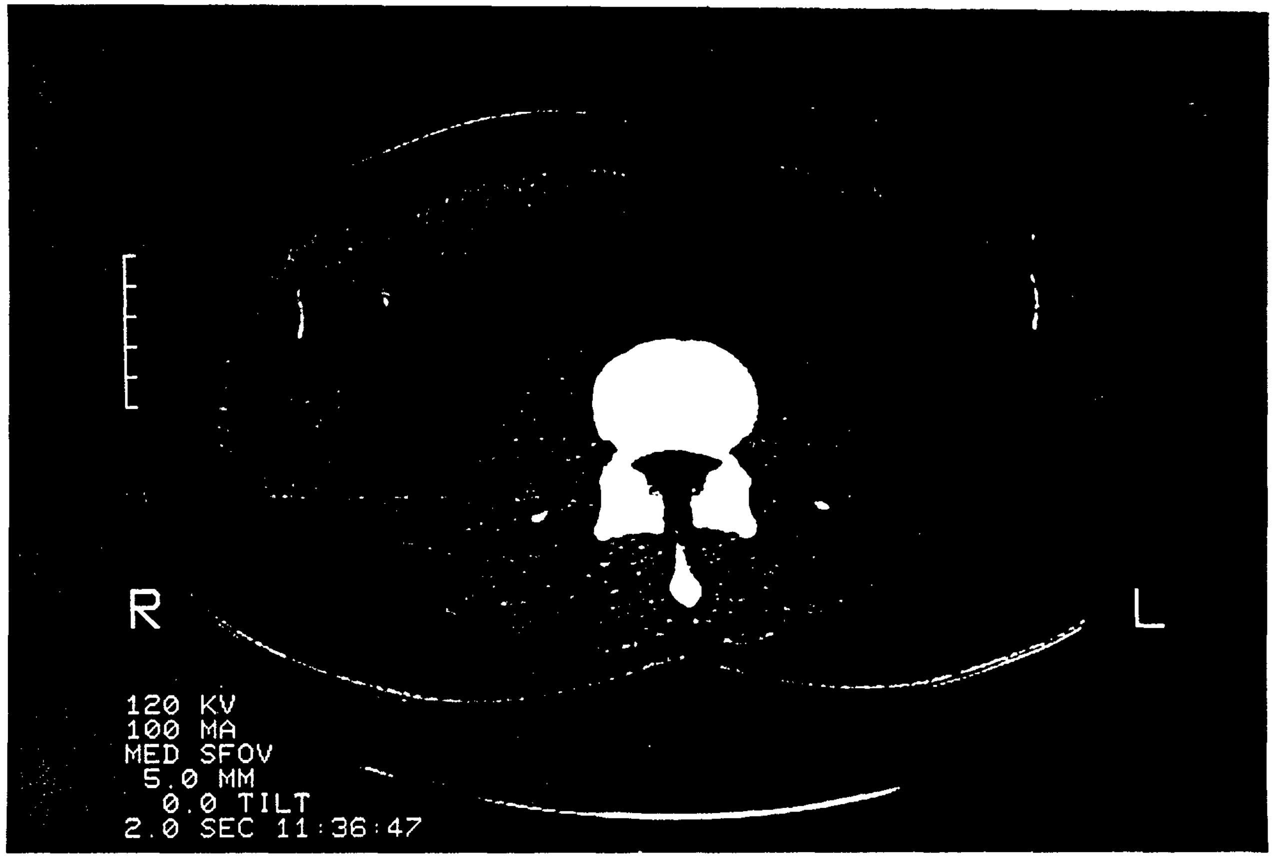

Figure 4.1. The CT-Scan image of the trunk at the L3/L4 level of the spine. Images of fine wire electrodes appear as bright white vertical lines originating from the right and left side of the fascial border of the abdominal muscle mass to the center of the internal oblique muscles. Fine wires are visible for both right and left internal oblique muscles.

Schmitz, 1992; Tan et al., 1993). The subject performed two trials of MVE of the trunk in an upright standing posture in four random directions: flexion, extension, and right and left axial rotation. The secondary axes were unlocked during all the trails. The subject was asked to reach to his/her MVE and hold it for 3 seconds. Two minutes of rest separated each test to avoid fatigue.

Step 3. Pure and combined exertion

Based on the MVE values, the computer randomly selected a test from combinations of 0, 30, 60, and 100% of MVE for flexion, extension, and rotation of the trunk. To simulate pure and combined exertions, visual feedback was given to the subjects. A monitor visually displayed trunk torque output and target torque in the plane of sagittal and axial torque. A target torque was set at 0, 30, 60, 100% of MVE trunk exertion along one or combination of two axes of the trunk. Each subject was instructed to match their output torque to the target torque for at least three seconds. To reduce the number of testing conditions and the risk of fatigue, only rotation to the right side is considered in this study. No load condition, zero condition along each axis, was repeated more than three times. However, these tests were part of the randomization process. The zero condition served as a means to calculate the baseline EMG for the normalization of the data. Based on the subject's performance verbal feedback was provided. The practice trial was continued until the subject felt comfortable with the task and indicated that he/she was ready to start the test.

4.2.4 Data analysis

4.2.4.1 Data Reduction

Based on Equation 1, the three seconds of each trial was selected where the subject performed closest to the target torque. The close distance was defined by low value of the control index as it was calculated by the following formula:

$$\text{Control Index} = \frac{\sqrt{\left(Sag_{Torque} - Sag_{Tag}\right)^2 + \left(Tran_{Torque} - Tran_{Tag}\right)^2}}{\sqrt{\left(Sag_{Tag}\right)^2 + \left(Tran_{Tag}\right)^2}} \tag{1}$$

Where :

Sag_{Torque} and $Tran_{Torque}$ = The generated sagittal (Sag) and transverse (Tran) torque

Sag_{Tag} and $Tran_{Tag}$ = The sagittal (Sag) and transverse (Tran) target torque

For all ten trunk muscles, the mean of the normalized Root Mean Square of the EMG signal (NRMS-EMG) was calculated for the selected 3 second window. Based on previous recommendations (Redfern, 1992), the time constant of 50 ms was used for RMS calculation. As suggested by Lavender et al. (1992) and Seroussi and Pope (1987), the EMG signal of each muscle was normalized to the maximum and rest value of the muscle. In this study, the highest mean NRMS-EMG values among all the trials was selected for the normalization procedure.

4.2.4.2 CT-SCAN

The CT-scan images were down loaded from the CT-scan unit into the optical disks. At the Radiology Department (School of Medicine, New York University, New York), the images were converted to 512x512 binary image files. Sigma Scan Pro image processing software (Jandel Scientific, San Rafael, CA) was used to process the CT-scan images. From the CT-scan images, the region of each trunk muscle and the width and depth of the subject's trunk was manually selected. Then, software calculated the cross-sectional areas, and distance of centroid of muscles from the center of the disc. The sagittal view of the spine was inspected to insure that the angle of L3/L4 vertebra with respect to the horizontal line was about 90° and that the values measured from these images do not need any correction, as suggested by McGill et al. (1993).

4.2.4.3 Calculation of Muscle Forces

Muscle forces and trunk moments were calculated based on the method suggested in Chapter III. A ridge regression method (Chapter III) was used to calculate the gain and intercept for the flexor and extensor trunk muscles. All calculations were performed for the L3/L4 level of the lumbar spine.

To investigate the effect of anatomical geometry on the results of the EMG-driven model, three geometrical values were used in the calculation of moments and spinal forces. Anatomical values and line of action for Geometry 1 is based on the values reported by Granata (1993). This data set is based on assumption that the IOB muscle acts as an extensor in the sagittal plane. Geometry 2 is similar to the first geometrical value, with a correction to the assumption made with respect to the value of IOB muscle. The value of the IOB lever arm is changed so that the IOB muscle is producing a flexion moment in sagittal plane. Geometry 3 represents the geometrical values of the trunk muscles as measured from the CT-scans of each in this study. The line of action muscles used in Geometry 3 was based on the study by Schultz et al. (1983).

4.2.5 Statistical Analysis

For each subject, the mean of absolute error (the difference between predicted and actual moment produced) of trunk moments, compression, and shear forces were calculated. Descriptive statistics (the mean and standard deviation) were computed for demographic data, absolute error of moments, compression and shear forces.

The statistical procedure MANOVA of repeated measure design (SAS User's Guide, 1985) was used to determine the effect of geometry on the results of the EMG-driven model of the load on the lumbar spine. If the overall multivariate analysis was significant, a univariate analysis (ANOVA) was performed. The Tukey test was performed to compare the results of the three anatomical models, to determine if the effect of the univariate test was statistically significant. In this study, statistical significance level was set at $P<0.05$.

4.3 Results

The CT-scan and EMG data of 6 subjects (4 males and 2 females) was used in the analysis. In addition to these subjects, the data from several subjects were collected but not included in the data analysis due to different technical reasons related to either inadequate CT-scan data (low resolution for separating the muscle or loss of calibration information of image), or missing EMG signal due to lost wire or surface electrode signal.

Mean age, weight, and height were 27.3 (±5.4) years, 68.0 (±6.9) Kg, and 169 (±8.8) cm, respectively. The scores for index of work, sport, and leisure were 2.40 (±0.66), 2.95 (± 1.68), and 3.25 (±0.57) respectively. Table 4.1 summarizes the mean and standard deviation of muscle parameters for six subjects.

For each subject, the mean of absolute error across all the trials was calculated. Table 4.2 shows the mean of absolute error of each plane, and mean of spinal forces for all subjects. The sagittal moment error was 30.2 ($\pm$ 6.7) Nm for Geometry 1, and 23.6 ($\pm$5.0) Nm and 18.2 ($\pm$ 4.5) Nm for Geometry 2 and 3 respectively. The results of the three geometries with respect to transverse and lateral error were very similar.

The results of Geometry 3 suggest higher anterior-posterior (AP) shear and compression forces compared to other geometries but lower lateral shear force. Geometry 2 indicated lower AP shear force and compression force compared to other geometries.

The results of the multivariate analysis of variance indicated significant ($p<0.001$) effect of Geometry values on the results of the EMG-driven model. The result of ANOVA indicated that sagittal error, AP and lateral shear forces, and estimated gains are significantly affected by the geometry (Tables 4.2 and 4.3).

The mean of absolute sagittal error for Geometry 1 was 30.2 ($\pm$ 6.7) Nm and it was significantly different than the other two geometries (see Tables 4.2 and 4.3). Also, the shear forces based on Geometry 1 were significantly ($p< 0.01$) different than the shear forces predicted based on the other geometries. The flexor gain predicted by Geometry 1 was significantly ($P<0.02$) different than gain predicted by Geometries 2 and 3. Significantly ($P<0.01$) higher extensor mean was predicated by the Geometry 3 compared to other two geometries.

Table 4.1. Mean and standard deviation of cross-sectional area and lever arm (the distance of the centroid of the muscle with respect to the center of the disc) for ten selected trunk muscles at the L3/L4 level of the spine.

	Lateral Lever Arm (mm)	Anterior-Posterior Lever Arm (mm)	Cross-Sectional Area (mm^2)
Left ES	37.32 (3.81)	59.54 (4.59)	2506.79 (443.61)
Right ES	35.49 (7.94)	59.17 (5.63)	2540.37 (400.2)
Left RA	41.91 (10.12)	74.33 (13.02)	855.81 (201.36)
Right RA	44.93 (3.0)	75.02 (15.81)	884.83 (168.17)
Left IOB	108.42 (11.13)	17.81 (8.06)	1458.91 (406.86)
Right IOB	110.78 (11.00)	19.32 (11.32)	1317.51 (335.85)
Left EOB	121.2 (10.83)	15.2 (10.06)	1335.25 (357.79)
Right EOB	122.29 (10.15)	16.19 (12.02)	1320.09 (239.49)
Left LAT	102.56 (11.61)	39.93 (11.83)	193.62 (54.46)
Right LAT	102.53 (10.01)	39.62 (14.99)	215.63 (52.97)

Table 4.2. The result of EMG-driven model of load on the trunk based on three geometrical values and univariate analysis of variance (ANOVA) for mean of absolute error of predicted moments and compression and shear forces.

	Geometry 1	*Geometry 2*	*Geometry 3*	*F value*
Sagittal Error (Nm)	30.2 (6.7)	23.6 (5.0)	18.2 (4.5)	0.006
Transverse Error (Nm)	17.3 (5.7)	17.7 (7.4)	15.6 (5.5)	N. S.
Lateral Error (Nm)	11.2 (5.0)	12.0 (4.7)	13.1 (3.7)	N. S.
Lateral Shear (N)	159.4 (62.6)	119.7 (36.9)	107.4 (36.1)	0.0204
AP Shear (N)	537.6 (275.8)	371.7 (147.0)	904.9 (176.5)	0.0001
Compression (N)	1723.5 (796.2)	1306.5 (496.2)	1888.7 (860.9)	N. S.
Flexors Gain (N/Cm^2)	1.18 (0.57)	0.81 (0.32)	0.66 (0.3)	0.0105
Extensor Gain (N/Cm^2)	0.31 (0.09)	0.36 (0.12)	0.63 (0.3)	0.0110
Flexors Intercept (Nm)	0.025 (0.04)	-0.02 (0.02)	-0.02 (0.2)	N. S.
Extensor Intercept (Nm)	-0.03 (0.06)	0.03 (0.06)	-0.16 (0.33)	N. S.

Table 4.3. The significance level of the Tukey multiple comparison test for main effect geometry on absolute error of predicted moments and compression and shear forces of the trunk.

	Geometry 1-2	Geometry 1-3	Geometry 2-3
Sagittal Error	1>2 0.04	1>3 0.001	N.S.
Lateral Shear	1>2 0.0311	1>3 0.0083	N.S.
A/P Shear	1>2 0.042	1<3 0.0004	1<3 0.0001
Flexion Gain	1>2 0.0256	1>3 0.0037	N.S.
Extension Gain	N.S.-	1<3 0.0049	2<3 0.0149

4.4 Discussion

Table 4.1 shows that the IOB and EOB muscles have both a large cross-sectional area and lever arm. This suggests that the oblique abdominal muscles can provide a large moment along the trunk axis. One can assume that biomechanical models of the trunk should be sensitive to the moments produced by the oblique muscles.

EMG-driven models of the trunk have been used to estimate the moments produced by the trunk muscles (Granata, 1993; Marras and Mirka, 1992). These models are sensitive to both the geometrical description of muscles and the amplitude of EMG activity. The anatomical geometry of muscles in previous studies were either measured from CT-scan and or MRI images, or from reported values in the literature. The classical references for trunk muscles geometry are Dumas et al., (1991), Tracy et al. (1989), and

Schultz et al. (1983). Dumas et al. (1991) reported line of action, cross-sectional area, and lever arm of trunk muscles based on a study of 3 cadavers. Tracy et al. (1989) reported cross-sectional area and lever arm of trunk muscles based on a study of 25 cadavers. However, some values reported in their study are based on a smaller number of specimens, for example, the LAT muscle geometrical values at level of L3/L4 is based on a single cadaver.

The validity of the EMG signal depends on the reliability of verifying the muscle site, the available space for electrode placement, the distance of this landmark from other muscles, and other issues that effect the quality of the EMG signal, i.e., skin preparation and underlying fat. There is a consensus with respect to the electrode placement for the EOB, RA, ES, and LAT (Granata, 1993; Marras and Mirka, 1992; Schultz; 1983). However, two sites have been suggested in the literature for monitoring the myoelectrical activity of the IOB muscles: a) the anterior triangle (Floyd and Silver, 1950), and b) the posterior triangle known as the lumbar triangle (Marras and Mirka, 1992; Marras and Mirka, 1990; Mirka and Marras, 1993). Both of these triangles represent an area of the IOB muscle that is not covered by the EOB muscle. The myoelectrical activity of IOB muscle is therefore suggested to be monitored by surface electrodes from these sites.

The anterior triangle is defined as the area between the inguinal ligament, a line from the anterior superior iliac spine to the umbilicus and midline of abdomen. This site was suggested by Floyd and Silver (1950) and has previously been used by several researchers (Basmajian and De Luca, 1985; Snijders et al., 1995). The lumbar triangle is the area between iliac crest, posterior end of the EOB muscle, and the lateral border of

LAT muscle (O'Rahilly, 1986; Williams et al., 1989). The studies that used the lumbar triangle reported that myoelectrical activity of the IOB muscle at this site indicated that the muscle acts as an extensor, and combination of the IOB moments to total trunk moments should be adjusted accordingly. A careful inspection of the model formulation and anatomical values used in these studies suggests that the extensor-like contribution of IOB muscles achieved by assuming the lever arm of the muscle is on the posterior side of the trunk, and therefore it takes the same sign as the extensors muscle (Granata 1993, Marras and Mirka, 1992).

An inspection of the CT-scan images gathered in this study raises concern with respect to the use of the lumbar triangle as a site for electrical activity of the IOB muscles. According to anatomical textbooks (O'Rahilly, 1986; Williams et al., 1989) the lumbar triangle should be similar to that shown in Figure 4.1. In this study, however, in most cases the textbook definition of the lumbar triangle at L3/L4 was not correct. In some cases, the posterior end of the IOB and EOB muscles at the L3/L4 level end together, see Figure 4.2. In other cases the triangle was covered by other muscle, see Figure 4.3. Such anatomical variation suggests that this landmark is not reliable to monitor electrical activity of IOB muscles.

No information is available in the literature with respect to the size of the lumbar triangle. Our dissection of one cadaver suggests that the size of the lumbar triangle in an average sized man is too small for placements of a pair of electrodes 2 cm apart, as recommended in the literature (Basmajian and De Luca, 1985), (see Figure 4.4).

Generally the lumbar triangle is covered with layer of fat. Inspection of CT-scan images indicates that any electrode placed on the skin above the lumbar triangle would be more than 2.5 cm from the muscle. Furthermore, the electrode would be at a relatively equal distance from the EOB, ES, and LAT muscle. This explains the cross-talk between EMG activity of the EOB and ES muscles, as observed in previous studies using this landmark (Marras and Mirka, 1990; Marras and Mirka, 1992).

To demonstrate the sensitivity of EMG-driven models to the anatomical geometry of the trunk muscles in general and, more specifically, to the IOB muscle, the results of three anatomical data sets were compared: 1) an anatomical description of trunk muscle as by Granata (1993), in which the internal oblique was considered to be extensor, 2) using the same anatomical data as described by Granata (1993) with a modification to the lever arm of the IOB muscles under the assumption that the muscle acts as a trunk flexor; and 3) using the anatomical description of the trunk muscles as they were measured from CT-scan images of subjects and the assumption that IOB muscles is functioning as a flexor muscle in sagittal plane..

Tables 4.2 and 4.3 demonstrate that the EMG-driven models of spine are sensitive to the geometrical description of trunk muscles and, more specifically, to the assumption made with respect to IOB muscles. The variation between the results of Geometries 1 and 2 in Table 4.2 is solely a reflection of the assumptions made with respect to the role of the IOB muscle in the sagittal plane. The result of Geometry 1 indicates a higher mean of absolute error for the sagittal plane. The means of absolute error for the lateral and transverse planes were not significantly different, since the variation in anatomical

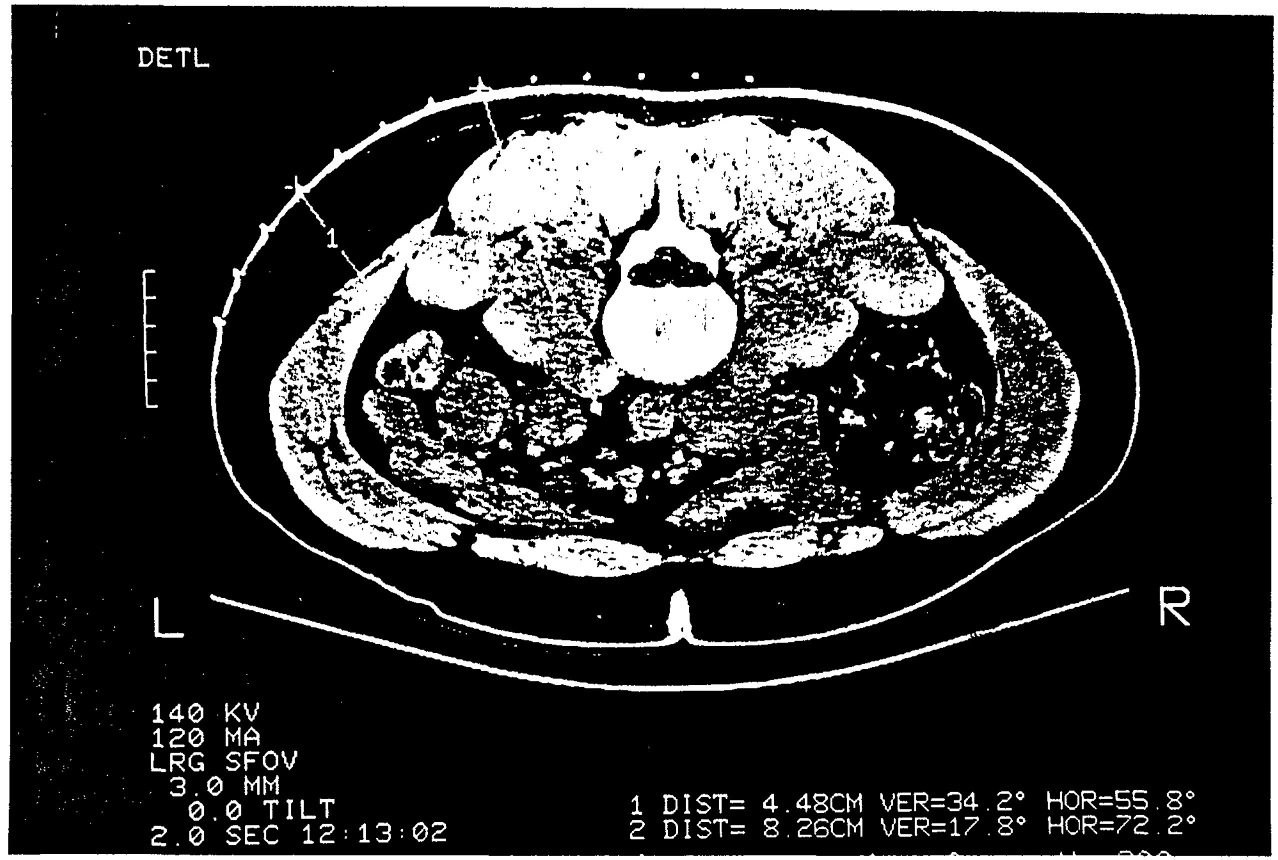

Figure 4.2. CT-scan image of the trunk at the L3/L4 level of spine. Both left and right internal oblique and external oblique muscles are ending together.

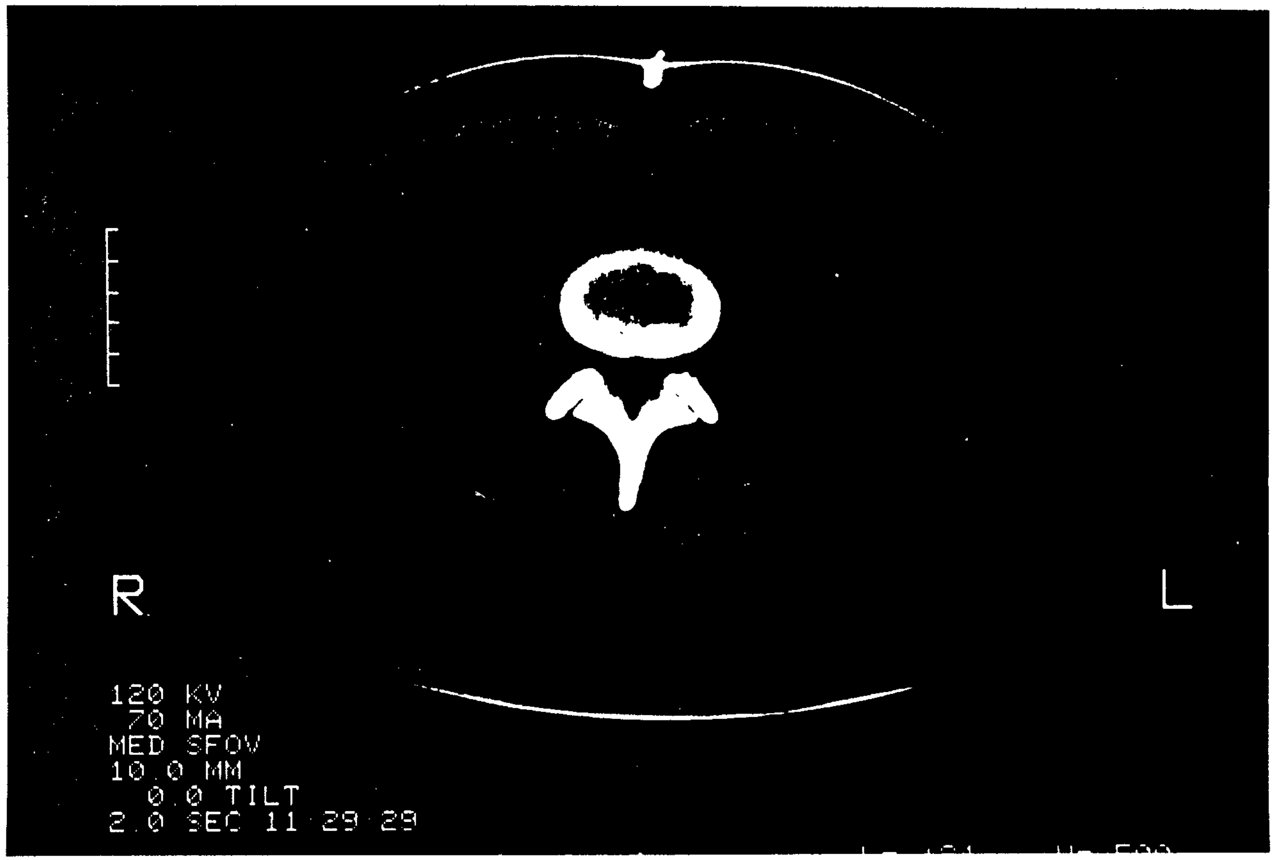

Figure 4.3. CT-scan image of the lumbar spine at the L3/L4 level of the spine. The lumbar triangle at left side is covered by the latissmius dorsi muscle.

Figure 4.4. Image of the lumbar triangle with a surface electrode shown over the triangle.

geometry was only with respect to the sagittal plane. The gain for the trunk flexor muscles based on Geometry 1 is higher than the values reported in the literature. .

The difference between the results of Geometries 2 and 3 is mainly the reflection of the sensitivity of the EMG-driven models to the accuracy of the anatomical data and the method of measurement of these values. A review of the literature suggests a low correlation among anthropometric variables and lever arm and cross-sectional area of the trunk muscles. Any error in the prediction of these values will directly effect the moment generated by the muscle and ultimately the net trunk moments, spinal forces, and gain calculated by a model.

The existing method of calculating the cross-sectional area and lever arm of the trunk muscles is only sensitive to the width and depth of the trunk. Variations among subjects in the trunk width and depth can affect the muscle cross-sectional area and lever arms. Therefore, one can not assume that two subjects with the same trunk size, an athletic individual (well-developed muscles and minimum skin fat) and a sedentary individual (small muscle mass and high skin fat), will have the same cross-sectional area and lever arm for the trunk muscles. Since these two individuals provide different levels of moments along each axis, the EMG-driven model will result in either a prediction of high gain value for the athletic individual or an underestimation of the moments and shear forces. This hypothetical case will demonstrate the shortcomings of statistically based geometry calculations and suggest a method for increasing the reliability of these values for use in biomechanical models.

Geometrical data collected from CT-scan or MRI images are more accurate in this respect. However, many other issues might affect the accuracy of the geometrical values gathered by these imaging techniques. In this study as well as other studies with CT-scan (Tracy et al., 1989) or MRI (Guzik et al; 1996), it is not always easy to distinguish the IOB, EOB, and LAT muscles at a the posterior site.

In most cases, the data gathered from these imaging techniques need to be corrected for viewing angle, as was suggested by McGill et al. (1993). In this study, we found the image with respect to the L3/L4 level to be very small (less than 3 degrees), and therefore no correction was made to the cross-sectional area and lever arm of the muscles. This is similar to the findings of Tracy et al. (1989), that reported -1.3 (±3.3) degrees for the L3/L4 level angle.

The cross-sectional area and lever arm of the muscles change significantly at the level of analysis with respect to the spine. Previous studies report a significant shift of centroid of the trunk muscle at different levels of the spine. In this study, we observed situations in which the cross-sectional area of the RA muscle for a subject was very small at the L3/L4 level, and a few millimeter shift of the level of CT-scan image along the longitudinal axis of the L3/L4 vertebrae trunk provide a reasonable cross-sectional area for this muscle. The only explanation for this observation was that the image at L3/L4 level provided information about the tendinous intersection of the RA muscle. These issues, as well as others, such as the posture selected for imaging (McGill et al., 1996), need to be standardized in order to insure the accuracy of the data collected by these imaging techniques.

4.5 Conclusion

The cost of imaging techniques such as CT-scan and MRI may not be justified for further biomechanical studies. However, at this early stage of biomechanical modeling, these imaging techniques are the only means of gathering information and formulating accurate assumptions about the anatomical geometry of the human body.

In an attempt to demonstrate this fact, we looked at the role of the IOB muscle in the formulation of the biomechanical models of the trunk and the assumptions made with respect to the role of this muscle. The result of this study shows the sensitivity of the sagittal moment, shear forces, and gains predicted by the EMG-driven models to the geometrical parameters of the trunk muscles used in these models.

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Chapter V

Conclusion and Future Studies

Epidemiological studies indicate that a combination of heavy external load and asymmetrical movement patterns cause a high internal load on the spinal structures and an increase in the risk of low back pain. Biomechanical studies indicate that the internal loads increase as the magnitude of external load increases or the net external load moves away from the sagittal plane and becomes an asymmetrical exertion. Summaries of these studies are presented in Chapter I. Two types of asymmetrical exertion can be assumed: asymmetrical exertion within a symmetrical posture and asymmetrical exertion within an asymmetrical posture. It is important to interpret the literature in light of the nature of asymmetrical exertion studied by the investigators.

In the prevention of low back pain and injury, it is essential to have a reliable estimate of risk for low back injury. This estimate requires clear understanding of the tasks and the individual's method of performance. Several methods were discussed in Chapter II and IV and the limitations of these methods were noted. The method and tools required for estimating the risk depends on: a) the reliability and validity required for the estimated risk, 2) the simplicity of gathering necessary information, and 3) the sensitivity of the model with respect to an error in the collection of information.

The EMG-driven models provide a better estimate of the internal forces for asymmetrical exertion of the trunk. Most of our knowledge regarding internal forces during asymmetrical exertion is based on studies that investigated internal spinal load during dynamic activities. The internal loads were estimated via EMG-driven models of the trunk. These studies provide a general understanding of human performance during asymmetrical posture. However, in interpreting the data provided by these studies, several concerns are raised in relation to the methodology and assumptions made in these models.

The biomechanical models are sensitive to anatomical data. Change in the centroid of the muscles, the lever arm, the cross-sectional area of the muscles, and the line of action of the muscle can directly affect the force produced by the muscle. The sum of these muscle forces provides information about total spinal force and the risk of injury. During dynamic trunk activity, the muscle parameters are continuously changing and ultimately there is currently no method of measuring or estimating these changes. Previous investigators used fixed values for these geometrical data and assumed that the movement through the range of motion would not affect these parameters. This is inconsistent with the existing information in the literature.

High spinal forces during the dynamic asymmetrical activity shown by these studies is partially due to a change of posture and thereby to changes in muscle alignments, and partially due to changes in net resultant moments. The aim of this study was to investigate the effect of changes in the magnitude and angle of the net resultant force on the EMG activity of trunk muscles and the internal spinal forces during upright pure and combined isometric trunk exertion.

The testing protocol of this study was based on a static, upright standing posture where only the magnitude and direction of net external trunk moments were manipulated. The direction of the net resultant moment was defined to simulate two symmetrical and three asymmetrical trunk exertions. The symmetrical exertions were pure sagittal exertion in the flexion and extension direction. The asymmetrical trunk exertions were simulated by exerting combined sagittal and transverse moments. The study demonstrated the effect of magnitude and direction of net resultant moment on the NRMS-EMG activity of ten trunk muscles. Three conclusions were drawn from the results of the data presented in Chapter III:

1. Ten selected trunk muscles were significantly affected by the magnitude and direction of the net resultant moment.
2. The NRMS-EMG of trunk muscles were higher during combined trunk exertion compared to pure exertion.
3. The controllability and capability of high magnitude trunk moments were significantly affected by the direction of the net resultant moment.

Chapters III and IV utilized the EMG-driven models to estimate the spinal force under pure and combined trunk exertion. An attempt was made to investigate the sensitivity of these models to the assumptions made with respect to geometrical and mathematical formulations.

Chapter III indicates that the results of the EMG-driven models of the trunk were significantly affected by the assumptions made with respect to the anatomical muscle parameters.

Chapter IV demonstrates that the spinal forces estimated by these models will significantly affect the assumptions related to the formulation of muscle force. Previous EMG-driven models assumed a linear EMG-force relationship. Furthermore, the intercept of this linear line was assumed to be zero. The shortcomings of this approach are discussed in Chapter IV. The results of spinal forces based on this approach were compared to other approaches where the intercept takes a non-zero value.

The new method of calculating the slope and intercept of the EMG-force relation has been discussed in Chapter IV. The main advantage of this method is the possibility of calculating the gain and intercept for each trunk muscle or a group of muscles with similar function.

The results of Chapters III and IV raise some concerns about the interpretation of the results of EMG-driven models. Previous studies presented the validity of their results by inspecting the value of estimated gain and the correlation coefficient of the predicted and measured moments. The correlation coefficient only provides information with respect to a similarity in patterns of predicted and measured moments. An error on the magnitude of predicted moments will not have an effect on the correlation coefficient. Chapter IV presents the results of three different methods of formulating EMG-driven models. All three methods estimated the gain value within the physiological range reported in the literature. However, significantly different spinal forces were estimated by these models.

Since internal forces on the spinal structures cannot be measured directly, there is no method for measuring the validity of one model compared to others. The results of the EMG-driven models should be interpreted based on: a) the validity of assumptions made in the formulation of the models, b) the accuracy of the predicted moments, c) the gain value estimated by the model, and d) that the spinal forces do not exceed the tissue's threshold.

5.1 Limitations and Future Research

The EMG-driven models assume that the calculation of spinal force at the level of L3/L4 vertebrae is the best representation of lumbar spine forces. In addition, these models assume that the EMG activity of the muscles and the forces estimated based on EMG activity is uniform throughout the muscle, and that EMG signals collected at the L3/L4 level are a reasonable representation of muscle activity. In our study as well as other studies, these assumptions are made based on practical limitations. In the future, more comprehensive data from the trunk muscle region should be collected and analyzed.

The results of this study cannot be generalized due to small sample size. Fifteen subjects were used in the first phase of the study (Chapter I), but we were only able to analyze 6 subjects in the second phase of the study (Chapters II and IV). Nine subjects were not included in the second phase due to technical reasons, including: a) in process of removing the needle, the wire came out by the needle rather than stay in the muscle, b) poor EMG signals were collected due to loss of contact, and c) the resolution of the CT-scan images was low and it was impossible to distinguish the muscle borders. Future work should focus on the improvement of these techniques to eliminate the risk of losing necessary information.

In this study, we assumed a linear EMG-force relationship. The effect of a nonlinear EMG-force relation should also be studied. The formulation of ridge regression in Chapter III can be expanded for this purpose.

EMG-driven models are sensitive to the anatomical parameters of the trunk muscles. The subject-specific muscle parameters should always be used in EMG-driven models. CT-scan images were used in this study to measure the cross-sectional area and lever arm of the ten trunk muscles. This is a relatively expensive method that has limitations in its use and availability. Due to the low resolution of the CT-scan images, image processing techniques fail to select the region of muscle properly. The region of muscles needs to be manually selected. If possible, a less expensive and simpler technique should be used in the future to measure these parameters.

We selected the line action of muscle based on the data available in the literature. There is no method of inspecting the validity of the existing data. One can assume that the results of EMG-driven models are sensitive to the accuracy of these values. The current information is based on a limited number of cadavers. The validity of this data and its possible effects on the EMG-driven models should be evaluated in the future.

Our study was limited to isometric, upright trunk exertion. The main concern in our study was to eliminate the possible errors due to a change of muscle alignment. We limited our study to ten selected trunk muscles and did not include all trunk muscles. Future work should concentrate on the refinement of these models and the improvement of methods used for acquiring data and estimating forces under dynamic conditions.

5.2 Application

Epidemiological studies have identified several risk factors associated with LBP. This association was established based on statistical methods. The EMG-driven models provide a means to investigate the cause and effect relationship between these risk factors and low back injury.

The present study provides both theoretical and methodological improvements for EMG-driven models. The refinements of these models enhances our understanding of human performance. Based on these models, prevention techniques can be designed to reduce the risk of low back injury during manual material handling and other demanding tasks for the back.

In clinical settings, the EMG-driven models provide a tool for investigating the effect of clinical or rehabilitation guidelines based on the physical condition of the patient. Based on the result of these models, new treatment techniques and exercises can be designed, and the efficacy of treatment protocols can be evaluated.

The results in Chapter II have special applications for ergonomic job evaluation and job design. It was demonstrated that controllability and capability of trunk exertion decreases as the task becomes asymmetrical. Furthermore, the capability of planar exertion cannot be used as an indication for the capability of the individual during combined exertion. This has implications for therapists and physical therapists, as well as for ergonomists who are involved in occupational rehabilitation (work hardening), return to work evaluation for patients, and job evaluation.

Appendices

Appendix A

Demographic and Medical Data Sheet

1. NAME OF PARTICIPANT: ______________________
2. I.D. NUMBER: ______________________________
3. DATE OF EXAMINATION: ______________________
4. AGE (in years): ___________________________
5. Date of Birth (mon. day, year):______________
6. WEIGHT (in kilogram): ______________________
7. HEIGHT (in centimeter): _____________________
8. HAND DOMINANCE: Right ______ Left _____
9. OCCUPATION: (Primary) ______________________

   (Other) ______________________

10. EMPLOYMENT OR OCCUPATION STATUS:

_____ Employed or Working Full Time

_____ Employed or Working Part Time

_____ Temporarily Employed or Working

_____ Not Employed or Working

_____ Retired

11. HIGHEST EDUCATIONAL LEVEL YOU HAVE COMPLETED:

_____ Grade School

_____ Some High School

_____ High School Graduate

_____ Some College

_____ College Graduate

_____ Some Graduate School

_____ Graduate Degree

Name: ____________________________

12. MEDICAL HISTORY:

a) What is your general health?

_____ Excellent
_____ Very good
_____ Good
_____ Fair
_____ Poor

b) Do you smoke?

YES ____ NO ____
If yes, how much per day? __________________

c) Do you suffer from any of the following?
(Check all items that applies to you).

	YES	NO
Arthritis	_____	_____
Thin or weak bones	_____	_____
Heart Disease	_____	_____
Heart Attack	_____	_____
Chest pain	_____	_____
High blood pressure	_____	_____
Diabetes	_____	_____
Hernia	_____	_____
Peptic ulcer	_____	_____
Frequent or recurrent headaches	_____	_____
Dizziness or vertigo	_____	_____
Ringing in the ear	_____	_____
Fainting spells	_____	_____
Loss of consciousness	_____	_____
Convulsions or seizures	_____	_____
Double vision	_____	_____
Loss of visual field(s)	_____	_____
Numbness or tingling sensations	_____	_____
Difficulty in breathing	_____	_____
Paralysis	_____	_____
Uncontrollable movements	_____	_____
Difficulty in walking	_____	_____
Difficulty in speaking	_____	_____
Loss of bowel control	_____	_____
Loss of bladder control	_____	_____

Name: ____________________________

Other medical condition(s) Yes____ No____
If yes, please specify ______________________
____
____

d) Are you on any medication?
YES ____ NO ____
If yes, please specify medication and reason for taking it ______________________________
____

e) In the past six months, did you have any low back pain significant enough to interfere with your work or daily activities?
YES ____ NO ____
If yes, when? ____

f) Do you have any history of low back injury or low back surgery?
YES ____ NO ____
If yes, please specify date and type of injury or surgery ___
____

g) Did you ever have any whiplash injury?
YES ____ NO ____
If yes, please specify date and type of injury ____
____

h) Have you been hospitalized for broken bones?
YES ____ NO ____
If yes, please specify part of the body and date of injury ______________________________
____

i) Do you have any type of muscle weakness?
YES ____ NO ____
If yes, how can you describe it? ____
____

j) Do you suffer from any neurological

k) Do you have any family history of neurological or muscular problems?
YES ____ NO ____
If yes, please specify ____
____

Name: ______________________________

l) Did you ever have injury or operations or both on any of these body parts?
YES ____ NO ____
If yes, please specify the date, the extent, and the presence of any residual problems.

	Date	Extent	Residual Problems
Lower limbs	__________	__________	__________
		__________	__________
		__________	__________
Upper limbs	__________	__________	__________
		__________	__________
		__________	__________
Neck	__________	__________	__________
		__________	__________
		__________	__________
Back	__________	__________	__________
		__________	__________
		__________	__________
Head	__________	__________	__________
		__________	__________
		__________	__________

m) Are you aware of any other physical condition(s) that may affect your safe completion of this study?
YES ____ NO ____
If yes, please specify ____
____

APPENDIX B

HABITUAL PHYSICAL ACTIVITY QUESTIONNAIRE

Name:______________________

Date:______________________

Main Occupation: ______________

Diredtions: On the blank space preceding each question, please write the number corresponding to the best answer. Some of the questions are repetitious for good reasons. Please bear with us and answer as best as you can.

_______ Wl. THE PHYSICAL ACTIVITY LEVEL OF MY MAIN OCCUPATION IS:
1) Low (i.e., clerical work, shopkeeping, driving, teaching, studying, housework, medical practice, and all other occupations with a university education)
2) Medium (i.e., factory work, carpentry, plumbing, and farming)
3) High (i.e., dock work, construction work, and sport)

_______W2. AT WORK I SIT:
1) never 2) seldom 3) sometimes 4) often 5) always

_______W3. AT WORK I stand:
1) never 2) seldom 3) sometimes 4) often 5) always

_______W4. AT WORK I WALK:
1) never 2) seldom 3) sometimes 4) often 5) always

_______ W5. AT WORK I LIFT HEAVY LOADS:
1) never 2) seldom 3) sometimes 4) often 5) always

_______ W6. AT WORK I kM TIRED:
1) always 2) often 3) sometimes 4) seldom 5) never

_______ W7. AT WORK I SWEAT:
1) always 2) often 3) sometimes 4) seldom 5) never

_______ W8. IN COMPARISON WITH OTHERS OF MY OWN AGE, I THINK MY WORK IS PHYSICALLY:
1) very often 2) often 3) sometimes 4) heavier 5) much heavier

_______ SI. DO YOU PLAY A SPORT?
1) Yes 2) No

IF YOU PLAY A SPORT:
WHAT IS YOUR PRIMARY OR MOST FREQUENTLY PLAYED SPORT?

____

_______ THE PHYSICAL INTENSITY LEVEL OF MY PRIMARY SPORT IS:
1) low (e.g., billiards, sailing, bowling, and golf)
2) medium (e.g., badminton, cycling, dancing, swimming, tennis, and middle distance running)
3) high'(e.g., boxing, basketball, football, rugby, rowing, and long distance running)

_______ IN A WEEK, I PLAY MY PRIMARY SPORT:
1) less than 1 hour
2) 1-2 hours
3) 2-3 hours
4) 3-4 hours
5) more that 4 hours

_______ IN A YEAR, I PLAY MY PRIMARY SPORT:
1) less than 1 month
2) 1-3 months
3) 4-6 months
4) 7-9 months
5) more than 9 months

IF YOU PLAY A SECOND SPORT:
WHAT IS YOUR SECOND SPORT? ______________________________

_______ THE PHYSICAL INTENSITY LEVEL OF MY SECOND SPORT IS:
1) low (e.g., billiards, sailing, bowling, and golf)
2) medium (e.g., badminton, cycling, dancing, swimming, tennis, and middle distance running)
3) high'(e.g., boxing, basketball, football, rugby, rowing, and long distance running)

_______ IN A WEEK, I PLAY MY SECOND SPORT:
1) less than 1 hour
2) 1-2 hours
3) 2-3 hours
4) 3-4 hours
5) more that 4 hours

_______ IN A YEAR, I PLAY MY SECOND SPORT:
1) less than 1 month
2) 1-3 months
3) 4-6 months
4) 7-9 months
5) more than 9 months

_______S2. IN COMPARISON WITH OTHERS OF MY OWN AGE I THINK MY PHYSICAL ACTIVITY DURING MY LEISURE TIME IS:
1) much less 2) less 3) the same 4) more 5) much more

_______S3. DURING MY LEISURE TIME I SWEAT:
1) very often 2) often 3) sometimes 4) heavier 5) much heavier

_______S4. DURING MY LEISURE TIME I PLAY A SPORT:
1) never 2) seldom 3) sometimes 4) often 5) very often

_______L1. DURING MY LEISURE TIME I WATCH TELEVISION:
1) never 2) seldom 3) sometimes 4) often 5) very often

_______L2. DURING MY LEISURE TIME I WALK:
1) never 2) seldom 3) sometimes 4) often 5) very often

_______L3. DURING MY LEISURE TIME I CYCLE (INCLUDING STATIONARY CYCLING):
1) never 2) seldom 3) sometimes 4) often 5) very often

_______L4. HOW MANY MINUTES DO YOU WALK/OR CYCLE PER DAY TO AND FROM WORK, SCHOOL AND SHOPPING?
1) less than 5 minutes
2) 5-15 minutes
3) 16-30 minutes
4) 31-45 minutes
5) more than 45 minutes

APPENDIX C

HABITUAL PHYSICAL ACTIVITY QUESTIONNAIRE SCORE CALCULATION

The following calculations are based on Baecke et al. (1982):

I. Calculation of scores of the work index of physical activity:

$$\text{Work Index} = \frac{W1 + (6-W2) + W3 + W4 + W5 + W6 + W7 + W8}{8}$$

where:

Wl = 1 if answer to question Wl is 1
= 3 if answer to question Wl is 2
= 5 if answer to question Wl is 3

W2, W3, W4 , W5 = 1 if answer is 1
= 2 if answer is 2
= 3 if answer is 3
= 4 if answer is 4
= 5 if answer is 5

W6, W7, W8, = 5 if answer is 1
= 4 if answer is 2
= 3 if answer is 3
= 2 if answer is 4
= 1 if answer is 5

II. Calculation of sport index of physical activity:

$$\text{Sport Index} = \frac{S1s + S2 + S3 + s4}{4}$$

Where:

$$S1s = \sum_{i=1}^{2} (\text{Intensity} \times \text{Time} \times \text{Proportion})$$

To determine the intensity equivalent score (S1a or S1b), the following numbers was substituted for the corresponding test scores:

0.76	for intensity score of 1 (i.e., low intensity sports);
1.26	for intensity score of 2 (i.e., medium intensity sports); or
1.76	for intensity score of 3 (i.e., high intensity sports).

To determine the weekly frequency equivalent score (Wa or Wb), the following numbers was substituted for the corresponding test scores:

0.5	for weekly frequency score of 1 (i.e., less .than 1 hour);
1.5	for weekly frequency score of 2 (i.e., 1-2 hours);
2.5	for weekly frequency score of 3 (i.e., 2-3-hours);
3.5	for weekly frequency-score of 4 (i.e., 3-4 hours); or
4-.5	for weekly frequency score of 5 (i.e., > 4 hours).

To determine the yearly frequency equivalent score, the following numbers was substituted for the corresponding test scores:

0.04	for yearly frequency score of 1 (i.e., less than 1 month);
0.17	for yearly frequency score of 2 (i.e., 1-3 months);
0.42	for yearly frequency score of 3 (i.e., 4-6 months);
0.67	for yearly frequency score of 4 (i.e., 7-9 months); or
0.92	for yearly frequency-score of 5 (i.e., > 9 months).

If the subject do not play a sport (.i.e., those who answered (2) to item SI) the Slc score is zero.

Value of S2 and S3 will be calculated similar to W3, and S4 will calculated similar to W6.

III. Calculation of Leisure index indices of physical activity:

$$\text{Leisure Index} = \frac{(6 - L1) + L2 + L3 + L4}{4}$$

Where L1 to L4 scores will calculated similar to W3 and W4.

Appendix D

Mean and Standard Deviation of EMG-RMS Data

Table D.1. Means and standard deviations for NRMS-EMG of left and right EOB muscle for three levels (30, 60, and 100 % MVE) and angles of net resultant moment (-45, 0, 45, 90, 135, 225) exertion.

		Right			Left		
		30%	60%	100%	30%	60%	100%
-45	Mean	.		85.87	.		72.77
	SD	.		24.87	.		20.9
	n	.		15	.		15
0	Mean	17.12	45.05	82.24	15.88	39.79	75.89
	SD	9.08	19.05	26.48	8.96	16.35	26.71
	n	14	15	15	15	15	15
45	Mean	14.68	46.64	70.89	19.55	64.85	87.88
	SD	8.75	19.57	24.89	10.31	21.64	19.67
	n	15	14	15	15	15	15
90	Mean	7.74	18.96	30.16	8.46	24.75	47.02
	SD	7.39	14.23	17.65	6.27	15.14	25.31
	n	15	15	14	15	15	15
135	Mean	6.8	20.88	31.36	6.62	22.48	35.24
	SD	4.35	13.32	11.69	4.51	15.72	14.36
	n	15	15	14	15	15	15
180	Mean	3.48	7.55	17.42	3.48	7.24	16.91
	SD	2.09	5.08	9.41	2.61	5.9	12.28
	n	15	15	14	15	15	15
225	Mean	.		45.34	.		31.72
	SD	.		23.58	.		21.97
	n	.		14	.		15

Table D.2. Means and standard deviations for NRMS-EMG of left and right IOB muscle for three levels (30, 60, and 100 % MVE) and angles of net resultant moment (-45, 0, 45, 90, 135, 225) exertion.

		Right			Left		
		30	60	100	30	60	100
-45	Mean	.		56.31	.		86.26
	SD	.		24.18	.		19.93
	n	.		12	.		11
0	Mean	14.6	45.59	62.47	15.03	28.95	62.72
	SD	12.15	20.16	21.99	7.59	13.38	24.61
	n	12	11	12	11	13	12
45	Mean	17.7	53.49	80.63	10.78	37.87	53.71
	SD	7.01	20.12	18.74	7.78	17.9	26.79
	n	12	12	12	13	12	13
90	Mean	15.71	36.64	74.29	7.1	24.23	39.63
	SD	16.12	17.17	25.67	9.1	15.2	18.72
	n	12	12	11	13	12	12
135	Mean	13.4	31.53	54.04	4.65	14.81	26.71
	SD	10.21	14.54	20.01	3.53	11.01	16.4
	n	12	12	12	12	12	12
180	Mean	3.83	11.07	17.93	7.17	11.75	23.58
	SD	5.26	15.44	10.77	5.28	12	15.41
	n	12	12	12	13	12	12
225	Mean	.		31.33	.		68.45
	SD	.		21.8	.		28.22
	n	.		12	.		12

Table D.3. Means and standard deviations for NRMS-EMG of left and right RA muscle for three levels (30, 60, and 100 % MVE) and angles of net resultant moment (-45, 0, 45, 90, 135, 225) exertion.

		Right			Left		
		30	60	100	30	60	100
-45	Mean	.		88.22	.		81.3
	SD	.		14.82	.		19.33
	n	.		14	.		14
0	Mean	11.59	44.76	79.68	13.29	43.18	91.09
	SD	6.11	19.17	26.34	5.14	15.35	14.72
	n	15	14	14	15	15	15
45	Mean	12	53.96	77.54	13.35	49.54	73.14
	SD	6.34	24.53	21.48	6.65	14.87	22.28
	n	14	14	15	15	15	15
90	Mean	2.47	10.42	13.4	2.17	6.32	15.61
	SD	2.61	18.29	7.39	2.06	5.24	10.64
	n	14	15	15	15	15	15
135	Mean	4.75	6.75	9.88	1.72	5.2	8.81
	SD	11.04	5.44	5.44	1.7	4.3	4.84
	n	15	13	15	15	15	15
180	Mean	2.09	2.87	6.04	0.91	2.56	5.51
	SD	3.7	3.26	6.05	0.72	3	3.77
	n	15	15	14	15	15	15
225	Mean	.		8.41	.		10.8
	SD	.		5.11	.		4.99
	n	.		15	.		15

Table D.4. *Means and standard deviations for NRMS-EMG of left and right ES muscle for three levels (30, 60, and 100 % MVE) and angles of net resultant moment (-45, 0, 45, 90, 135, 225) exertion.*

		Right			Left		
		30	60	100	30	60	100
-45	Mean	.		10.66	.		21.59
	SD	.		4.03	.		24.94
	n	.		15	.		15
0	Mean	1.09	3.01	11.78	1.32	10.25	14.86
	SD	1.94	2.05	12.1	2.02	25.98	23.9
	n	15	15	14	15	14	15
45	Mean	5.52	9.56	15.11	1.59	5.66	10.93
	SD	7.54	8.86	10.95	1.31	2.81	7.67
	n	15	15	15	15	15	15
90	Mean	11.78	18.14	27.05	9.98	14.84	24.19
	SD	5.56	10.99	11.44	5.59	11.52	17.52
	n	15	15	15	14	15	15
135	Mean	26	54.63	88.17	24.88	58.21	80.47
	SD	6.94	16.94	13.47	11.6	25.28	20.29
	n	15	15	15	15	14	14
180	Mean	28.46	46.36	77.05	24.35	46.72	73.26
	SD	8.82	10.82	16.47	10.98	19.1	23.88
	n	15	15	15	15	15	14
225	Mean	.		84.43	.		79.84
	SD	.		16.56	.		24.71
	n	.		15	.		14

Table D.5 Means and standard deviations for NRMS-EMG of left and right LAT muscle for three levels (30, 60, and 100 % MVE) and angles of net resultant moment (-45, 0, 45, 90, 135, 225) exertion.

		Right			Left		
		30	60	100	30	60	100
-45	Mean	.		9.67	.		71.7
	SD	.		19.65	.		28.54
	n	.		14	.		15
0	Mean	2.08	3.74	20.4	3.93	5.29	27.81
	SD	2.67	6.24	20.98	4.38	6.23	24.96
	n	14	14	14	15	15	15
45	Mean	12.77	32.84	66.09	2.85	7.06	10.78
	SD	11.81	23.17	29.16	5.09	11.18	15.22
	n	14	14	14	15	15	15
90	Mean	10.51	28.72	72.72	1.68	5.87	9.75
	SD	9.21	16.22	15.88	1.49	11.26	12.14
	n	14	14	14	15	15	15
135	Mean	12.25	40.12	75.97	5.46	17.41	30.44
	SD	8.68	21.14	29.27	6.34	24.05	25.31
	n	14	14	14	15	15	15
180	Mean	5.95	15.89	32.2	7.66	16.93	33.68
	SD	4.57	14.06	28.73	5.19	13.45	23.68
	n	14	14	14	15	15	15
225	Mean	.		27.1	.		81.85
	SD	.		29.3	.		24.74
	n	.		14	.		15

Appendix E

Consent form Hospital for Joint Diseases

ORTHOPAEDIC INSTITUTE
BERNARD ARONSON PLAZA, 301 EAST 17TH STREET
NEW YORK, NEW YORK 10003

CONSENT FOR PARTICIPATION IN SCIENTIFIC INVESTIGATIONS

NAME OF SUBJECT	NAME OF INVESTIGATOR
	Ali Sheikhzadeh
TITLE OF PROJECT	PROJECT NUMBER
Validity of surface electrode for monitoring EMG activity of the internal oblique muscle	22-103

INSTRUCTIONS

TYPE IN LAY TERMS THE NATURE AND PURPOSES OF THE STUDY, THE DURATION OF THE SUBJECT'S PARTICIPATION, THE BENEFITS, RISKS AND DISCOMFORTS OF THE PROCEDURES/TESTS AND/OR DRUGS/DEVICES TO BE USED, THE REQUIREMENTS (SAMPLE COLLECTIONS, BIOPSIES, QUESTIONNAIRES, ETC.) AND RESTRICTIONS (DIET, ETC.) AND DISCLOSE APPROPRIATE ALTERNATIVE TREATMENT. PROVIDE THE NAMES OF PERSONS TO CONTACT FOR QUESTIONS ABOUT THE STUDY AND IN CASE OF RESEARCH-RELATED INJURY.

I, ________________ agree to be a volunteer subject in the project entitled "Validity of surface electrode for monitoring EMG activity of the Internal Oblique muscle". I understand that 5 subjects will be studied at the Hospital for Joint Diseases over a four month period. All the subjects in the study must be males between the ages of 20–40 years old with no history of back pain for the last six months, and should be in good physical condition.

I understand that the experiment will be performed in four parts. First, I will fill out a questionnaire that requires some personal information and my medical history. All this information will be kept confidential.
Second, I understand I will undergo a CT-scan of my trunk at the Radiology Department of the Hospital for Joint Diseases. I understand that the exposure from CT-scan is comparable to half of a chest x-ray and which is considered as minimal risk of radiation exposure. Third, I understand that four wire electrodes and six disposable surface electrodes will be placed on selected sites of my trunk to record muscle activity. I understand a physician will insert the wire electrodes to my abdominal muscles by a hypodermic needles.

(Continue next page)

AUTHORIZATIONS

I, ____________________________, HAVE READ THE ABOVE AND HAVE BEEN GIVEN A CLEAR ORAL EXPLANATION OF THE NATURE, REQUIREMENTS AND EFFECTS OF THE STUDY AND SATISFACTORY ANSWERS TO MY INQUIRIES. I ACCEPT THE CONDITIONS OF THE STUDY AND I AUTHORIZE THE ABOVE PROCEDURES/TESTS TO BE PERFORMED AND/OR INVESTIGATIONAL DRUGS/DEVICES TO BE USED. I UNDERSTAND THAT IN THE EVENT OF PHYSICAL INJURY RESULTING FROM THIS STUDY, ONLY IMMEDIATE ESSENTIAL MEDICAL TREATMENT AS DETERMINED BY THE HOSPITAL WILL BE AVAILABLE FOR THE INJURY WITHOUT CHARGE TO ME PERSONALLY. THERE WILL BE NO MONETARY COMPENSATION. I ALSO REALIZE THAT I AM FREE TO WITHDRAW THIS CONSENT AT ANY TIME WITHOUT PREJUDICE TO MY FUTURE TREATMENT.
I UNDERSTAND THAT RECORDS OF THIS INVESTIGATION WILL BE KEPT CONFIDENTIAL BUT ARE SUBJECT TO INSPECTION BY THE U.S. FOOD AND DRUG ADMINISTRATION.

SIGNATURE OF PATIENT	DATE	SIGNATURE OF PERSON GIVING PERMISSION	DATE	VALID UNTIL
		RELATIONSHIP TO PATIENT:		FEB 14 1995

I, ____________________________, HAVE CLEARLY AND FULLY EXPLAINED TO THE ABOVE PATIENT (OR PERSON GIVING CONSENT) THE NATURE, REQUIREMENTS AND FORESEEABLE RISKS OF THE STUDY. IN MY JUDGEMENT HE/SHE IS FULLY COMPETENT TO COMPREHEND THE NATURE OF THE STUDY AND THE PROCEDURES INVOLVED.

SIGNATURE OF INVESTIGATOR	DATE	SIGNATURE OF AUDITOR-WITNESS	DATE	(STAMP)

BERNARD ARONSON PLAZA, 301 EAST 17TH STREET
NEW YORK, NEW YORK 10003

CONSENT FOR PARTICIPATION IN SCIENTIFIC INVESTIGATIONS

NAME OF SUBJECT	NAME OF INVESTIGATOR
	Ali Sheikhzadeh

TITLE OF PROJECT	PROJECT NUMBER
Validity of surface electrode for monitoring EMG activity of the internal oblique muscle	22-103

INSTRUCTIONS

TYPE IN LAY TERMS THE NATURE AND PURPOSES OF THE STUDY, THE DURATION OF THE SUBJECT'S PARTICIPATION, THE BENEFITS, RISKS AND DISCOMFORTS OF THE PROCEDURES/TESTS AND/OR DRUGS/DEVICES TO BE USED, THE REQUIREMENTS (SAMPLE COLLECTIONS, BIOPSIES, QUESTIONNAIRES, ETC.) AND RESTRICTIONS (DIET, ETC.) AND DISCLOSE APPROPRIATE ALTERNATIVE TREATMENT. PROVIDE THE NAMES OF PERSONS TO CONTACT FOR QUESTIONS ABOUT THE STUDY AND IN CASE OF RESEARCH-RELATED INJURY.

I understand that the wires will remain fixed in the muscles during testing. The wire will be removed by the physician at end of the testing. The pain involved is comparable with the pain involved in hypodermic injection. No other complication has been reported in the literature for this procedure. In order to verify the wire electrode placement, an ultrasound recording will be performed at certain area of my back. I understand there is no risk involve in ultrasound recording. Fourth. I will perform exertion tasks on the B200 Isostation. I will be asked to exert my maximumal effort against a constant resistance in trunk forward bending, backward bending, and twisting.

I understand that no medical benefits will accrue to me for my participation in this project, and that the data gathered will be useful in better understanding muscle activity of the trunk. I understand that a full effort will be made to answer any questions I have concerning the project. I understand that all information pertaining to me will remain confidential. I understand I have the right to withdraw from the study at any point during testing without giving any reason for doing so.

I understand that the Hospital for Joint diseases Orthopaedic Institute will provide immediate essential care for any injury and illnesses resulting from my participation in this study. However, neither long-term hospital treatment nor financial compensation will be available from the hospital. Further information may be received from my physician. If I have any questions, I can contact Ali Sheikhzadeh at (212)255-6690 or I will contact the Patient Representative of the Hospital for Joint Diseases, Ms. Pam Foster, at (212) 598-6474.

AUTHORIZATIONS

I, ______________________, HAVE READ THE ABOVE AND HAVE BEEN GIVEN A CLEAR ORAL EXPLANATION OF THE NATURE, REQUIREMENTS AND EFFECTS OF THE STUDY AND SATISFACTORY ANSWERS TO MY INQUIRIES. I ACCEPT THE CONDITIONS OF THE STUDY AND I AUTHORIZE THE ABOVE PROCEDURES/TESTS TO BE PERFORMED AND/OR INVESTIGATIONAL DRUGS/DEVICES TO BE USED. I UNDERSTAND THAT IN THE EVENT OF PHYSICAL INJURY RESULTING FROM THIS STUDY, ONLY IMMEDIATE ESSENTIAL MEDICAL TREATMENT AS DETERMINED BY THE HOSPITAL WILL BE AVAILABLE FOR THE INJURY WITHOUT CHARGE TO ME PERSONALLY. THERE WILL BE NO MONETARY COMPENSATION. I ALSO REALIZE THAT I AM FREE TO WITHDRAW THIS CONSENT AT ANY TIME WITHOUT PREJUDICE TO MY FUTURE TREATMENT.

I UNDERSTAND THAT RECORDS OF THIS INVESTIGATION WILL BE KEPT CONFIDENTIAL BUT ARE SUBJECT TO INSPECTION BY THE U.S. FOOD AND DRUG ADMINISTRATION.

SIGNATURE OF PATIENT	DATE	SIGNATURE OF PERSON GIVING PERMISSION / RELATIONSHIP TO PATIENT	DATE	VALID UNTIL

I, ______________________, HAVE CLEARLY AND FULLY EXPLAINED TO THE ABOVE PATIENT (OR PERSON GIVING CONSENT) THE NATURE, REQUIREMENTS AND FORESEEABLE RISKS OF THE STUDY. IN MY JUDGEMENT HE/SHE IS FULLY COMPETENT TO COMPREHEND THE NATURE OF THE STUDY AND THE PROCEDURES INVOLVED.

FEB 14 1995

SIGNATURE OF INVESTIGATOR	DATE	SIGNATURE OF AUDITOR-WITNESS	DATE

(STAMP)

DISTRIBUTION: AS MARKED ON COPIES. MAKE COPIES FOR SUBJECT (OR PERSON GIVING PERMISSION) AND PHARMACY (WHEN APPLICABLE)

Appendix F

Publication

TRUNK PERFORMANCE DURING COMBINED SAGITTAL AND TRANSVERSE ISOMETRIC EXERTION AT UPRIGHT STANDING

Ali Sheikhzadeh*, Mohamad Parnianpour†, Margareta Nordin*
*Occupational and Industrial Orthopaedic Center, Hospital for Joint Diseases Orthopeadic Institute, New York University Medical Center, 63 Downing St., New York, New York 10014
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Introduction: Maximum isometric trunk exertions are highly coupled especially during axial exertions. Since there is no muscle group with the primary task of axial rotation, several trunk muscles are recruited to generate torque or create motion in the transverse plane. Higher level of coactivity leads to higher internal loading of spine. This leads to higher strain and stress in the annulus fibers and on the facet joints, hence a higher risk of injury. The trunk muscles response to combined external sagittal and transverse moments is expected to lead to a higher loading than expected by the sum of responses to each individual moment. The purpose of this study was to test the hypothesis that the mean of RMS-EMG of ten selected trunk muscles will significantly change with the orientation of the net resultant moments about the trunk. In addition, the controllability during combined exertion were compared with planar exertion.

Methods: Ten males and 4 females with no prior history of low back pain in the past six months participated in the study. Mean age, height and were 30 (±7) years, 70 (±12) cm, and 168 (±9) Kg, respectively. A triaxial dynamometer (the B200 Isostation), was used to measure the torque generated during maximal and submaximal trunk exertion. In upright standing posture, the maximum isometric exertion (MVE) was measured for flexion, extension and right and left trunk rotation. To simulate pure and combined exertions, visual feedback was given to the subjects. A monitor displayed trunk torque output and target torque in the plane of sagittal and axial torque. A target torque was set at 0, 30, 60, 100% of MVE exertion along one or two axes of trunk. Subjects were instructed to match the their output torque to target torque for at least three seconds. A measure of controllability of the generated torque was computed by the distance from output torque to target torque. The lower value indicated a greater ability to

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generate and control the desired torque levels. Surface electrodes were used to monitor bilateral trunk muscles at L3 level for external oblique (EOB), rectus abdominis (RA), erector spinae (ES), and at L1-T12 level for latissimus dorsi (LD). Wire electrodes were used to measure bilateral muscle activity of internal oblique (IOB). Mean RMS-EMG and mean torques were computed for 3 seconds data where the subject performed closest to the target. A repeated MANOVA was performed to investigate the effect of the three levels of randomized exertion (30, 60 and 100% of MVE) and five angles of exertion (0, 45, 90, 135 , 180 degree) on ten trunk muscles. The five angles represent the combination of sagittal and axial torque where 0° indicates pure flexion and 180 is pure extension of the trunk.

__Results__: The amplitude of all muscles were significantly ($p<.001$) changed by the main effects of level and angle of exertion and their interaction. IOB, EOB and LD had significantly ($p<0.01$) higher activity during combined exertion compared to pure exertion. The controllability of the torque generation significantly reduced in the combined exertion ($p<0.01$). In addition, the maximum attainable toque under combined condition was less than the maximum pure voluntary capacity.

__Discussion__: Physically demanding tasks in industry are better represented with combined (coupled) task demands. Due to wide-spread availability of unaxial dynamometers, uniplanar maximum voluntary exertions have been used for quantification of trunk muscle strength. Based on present results, it is incorrect to assume a person's ability to perform the combined exertion based on exertions performed in one plane. The internal loading is higher for combined loading since the level and angle of exertion are significantly higher for all muscles. This also exposes the passive structure to a larger magnitude of load. The capability and controllability of the generated torques are significantly lowered during combined exertions. These results suggest the trunk muscles will be taxed at a higher relative load while performing combined loading tasks, increasing muscle fatigue and the probability of injury.

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