BIOMECHANICAL ANALYSIS OF THE PHYSICAL LOADS ON THE LOW BACK AND SHOULDER DURING DRYWALL INSTALLATION

BY

LU YUAN
B.E. BEIJING INSTITUTE OF TECHNOLOGY, CHINA (2000)
M.S. UNIVERSITY OF MASSACHUSETTS LOWELL (2004)

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Signature of Author: Lu Yuan **Date:** 9/14/06

Signature of Dissertation Supervisor: _____

Signatures of Other Dissertation Committee Members: _____

Laura Punnett

_____

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Dissertation Supervisor: Bryan Buchholz, Ph.D.
Professor, Department of Work Environment

ABSTRACT

This research provided a biomechanical analysis of the physical loads on the low back and shoulder during drywall installation. Both the ergonomic exposure assessment and intervention evaluation were conducted integrating the methods of work-sampling based observation, computer-aided simulation, biomechanical models of the low back and the shoulder, and a one-compartment challenge-recovery model.

In the first study, the Monte-Carlo simulation technique was used to generate input information for the biomechanical models from PATH (Posture, Activities, Tools, and Handling) observation data during a simulated 8-hour workday. We then estimated the required contraction forces of five paired trunk muscles; joint resultant forces of disc compression, lateral shear, and anterior-posterior shear at L4/L5; contraction forces of nineteen major shoulder muscles and one ligament; and resultant forces at the glenohumeral and sternoclavicular joints using the double linear optimization program and the nonlinear optimization program in MATLAB, respectively.

The second study examined trunk and shoulder muscle fatigue by introducing the concept of muscle force production capacity. The prevalences of muscle fatigue and consecutive fatigue periods were computed to illustrate the influence of muscle fatigue on physical capacity during drywall installation, where it was found that trunk muscles were more vulnerable to fatigue than shoulder muscles.

The third study evaluated the relative efficacy of three simulated ergonomic interventions for drywall installation. By comparing muscle contraction forces, joint resultant forces, and muscle fatigue parameters obtained from the integrated methods between post- and pre- interventions, it was determined that all of the proposed

interventions, including lifting assistance, two-person team work, and optimal work-rest schedule, contributed to the reduction of ergonomic hazards for drywall installers. The combination of lifting assistance and optimal work-rest schedule was the most efficacious intervention.

The research initiates a cost-effective protocol for physical ergonomic exposure assessment and intervention effectiveness evaluation, which can be applied to the construction industry and other non-routine work sectors where it is either infeasible or too labor/time intensive to explore direct measurement technology.

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF ILLUSTRATIONS

I. Introduction

A. Objective and Specific Aims

The objective of the research was to conduct a biomechanical analysis of the physical loads on the low back and shoulder during drywall installation in the commercial construction industry, based on information from field observations. In the Construction Occupational Health Program (COHP) in the Department of Work Environment at the University of Massachusetts Lowell, a number of hours of observations on drywall installation were made, from which both quantitative and qualitative data about ergonomic exposures were obtained. Specifically, the PATH methodology (Posture, Activity, Tools, and Handling), a work-sampling based approach, was applied to collect information about the frequency of ergonomic exposures. The goals of this project were to construct general biomechanical models to quantify the physical loads on the low back and shoulder for drywall installers, to define and identify the prevalences of muscle fatigue during a simulated "typical" workday, and to propose and evaluate intervention strategies to reduce the ergonomic hazards.

The specific aims of the research were to:

1. **Estimate muscle forces and joint resultant forces at the low back and shoulder based on PATH data:** The combination of biomechanical models and computer simulations was intended to provide a more comprehensive evaluation of the physical requirements on the low back and shoulder during drywall installation than that obtained from PATH alone. An 8-hour-workday activity series was firstly generated using the Monte-Carlo simulation method. Values of independent variables for biomechanical models of the low back and shoulder, such as anthropometric

parameters, joint angles in three dimensions, external load position and force vectors, and internal muscle unit force vectors and muscle generated moment vectors, were then estimated. Using the optimization programs in MATLAB, a numeric computation package, the required muscle forces and joint resultant forces during drywall installation were determined.

2. **Examine the cumulative effects of potential muscle fatigue using a one-compartment challenge-recovery model:** Muscle force production capacity was introduced to help better understand muscle fatigue during the simulated 8-hour drywall installation work. By comparing the required muscle contraction forces and muscle force production capacities, muscle fatigue was quantified.

3. **Determine the biomechanical advantages of different intervention strategies**: Several potential injury prevention strategies exist to reduce the ergonomic hazards and promote drywall installers' health and safety. These strategies include the use of assistive devices, team work applications, and appropriate rest periods. Despite the existence of these strategies, little is known about their efficacy and effectiveness in the construction workplace. Comparisons of the model output values between pre- and post- intervention, including muscle forces, joint resultant forces, and muscle fatigue, were estimated as proxies for the efficacy of possible interventions and provided guidelines for further study of their effectiveness.

B. Background and Significance

1. Physical ergonomic exposures and musculoskeletal outcomes during drywall installation

In the construction industry, drywall sheets are hung by drywall installers who are usually part of the carpenters' union in the U.S. (Pan et al., 1999). Drywall consists of gypsum formed into a flat sheet and sandwiched between two pieces of heavy paper. A typical drywall sheet weighs from 24.5 to 101.6 kg (54 to 224 lb) and is 1.22-m (4-feet) wide, 2.44- to 4.27-m (8- to 14-feet) long, and 9.5- to 25.4-mm (3/8- to 1-inches) thick. Drywall installation involves several steps. First, the drywall installers measure and cut the drywall sheets to fit the room dimensions and accommodate any openings such as electrical fixtures, windows, or doors. Then, the installers lift and carry each sheet to its destination and hold it in place while driving screws or nails to fasten the structure using powered screwdrivers or hammers.

To hang drywall onto the ceiling or the wall, drywall installers are constantly involved both in handling heavy and bulky materials and repetitive screwdriving motions. Moving drywall sheets from where they are stored to where they are being hung exposes drywall installers to potential ergonomic risk factors such as awkward postures and excessive use of force. Moreover, many characteristics of the construction industry, such as the unstructured worksite layout, confined spaces, and project deadline issues, may pose additional risks to the drywall installers during physical exertion. Well-documented injury events for drywall installers include overexertion, falls from heights, bodily reaction, and struck by an object (Hsiao and Stanevich, 1996; Chiou et al., 2000; Lemasters et al., 1998; Lipscomb et al., 1997 and 2000). The body parts most commonly

injured are the axial skeleton and shoulder, where back sprains, simultaneous sprains to the back and neck, and shoulder strains appear grievous.

Hsiao and Stanevich reported that, in 1987, drywall installers accounted for 1.41% (70,510) of all the workers in the U.S. construction industry and their compensable injury rate (27.5 cases per 100 workers) was three times the mean rate (9.5 cases per 100 workers) for the whole construction industry. Overexertion (28%), fall from height (24.7%) and being struck by an object (15.1%) were the predominant types of injury.

Chiou et al. (2000) analyzed the nonfatal traumatic injuries with days away from work among drywall installers using the Occupational Injury and Illness Survey conducted by the Bureau of Labor Statistics from 1992 to 1995. There were a total of 16,023 traumatic injuries for wage-and-salary drywall installers over this period but the rates decreased from 4,680 in 1992 to 3,065 in 1995. Half of the injuries resulted in at least 8 days absence and a quarter resulted in at least a month absence. More than 40% of the injured drywall installers suffered sprains, strains, and/or tears to muscles, tendons, ligaments or joints. The most frequently injured body part was the trunk, and more than 1/3 of the trunk injuries were most likely associated with handling solid building materials, especially drywall sheets.

Lemasters et al. (1998) computed the prevalence of work related musculoskeletal disorders among active union carpenters and found that the upper extremities, especially the shoulders, hands and wrists, were the most frequently injured body parts within the drywall or ceiling subspecialty. Drywall and ceiling installation entails repetitive, forceful use of screw guns often held in awkward positions, requiring sustained overhead reaching.

Lipscomb et al. (1997) examined the workers' compensation claims of musculoskeletal injuries and disorders among union carpenters in Washington State during the period 1989-1992. The analyses revealed that individuals who were mostly involved in drywall installation had rates of injuries to the axial skeleton (back sprains, back and neck sprains, and back ill-defined conditions) that were twice as high as those unionized counterparts doing heavy commercial carpentry work or pile driving. These individuals also had higher rates of sprains to the shoulder, forearm, knee, and ankle than other union carpenters.

Later, Lipscomb et al. (2000) examined health insurance eligibility files to investigate work-related injuries in drywall installation in western Washington State between 1989 and 1995. Of 1,773 drywall installers, 1,046 filed 2,567 claims. The most common mechanisms of injury consisted of being struck (38.3%), overexertion (28.1%), and falls (13.2%). Struck by injuries commonly resulted in cuts to the upper extremities. Overexertion injuries were most commonly described as sprains or strains involving the back, and sheetrock was associated with over 40 percent of these injuries. Falls most commonly produced injuries to the knees followed by the back and multiple injuries.

In summary, drywall installation involves significant exposure to ergonomic and safety hazards, especially overexertion with awkward body postures, which causes or aggravates physical injury, particularly at the low back and shoulder.

2. Ergonomic exposure assessment

Theoretically, there are a variety of ergonomic assessment methods and instruments available for evaluating hazardous tasks and activities. Several issues

including exposure variability, load tolerance, and muscle fatigue need to be discussed during ergonomic exposure assessment.

2.1. Work sampling approach

Over the past decade, researchers in COHP have been performing scientific research to prevent injury and illness among construction workers. With the goal of quantifying ergonomic exposures, COHP members developed PATH (Posture, Activity, Tools, and Handling), an observational work-sampling based technique in which the worker's tasks and activities, body postures, tools used, and loads handled are recorded at short fixed intervals (Buchholz et al., 1996).

PATH is used to determine the frequencies of activities conducted, postures of body parts assumed, and weights of material handled. The percentages of time that workers are exposed to different activities, awkward postures of the trunk, shoulder, and leg, and different loads in the hands, can be computed to reflect the real work. Since PATH data are collected based on the principle of a snapshot, they may also provide some information about the relative duration of each single activity. In that scenario, activities that take longer time are likely to be captured more frequently than those which require a shorter time to conduct.

Based on the comparisons between PATH and direct measurement systems, Paquet et al. (2001) demonstrated the validity of PATH. As an observational approach, PATH does not provide discrete values of the independent variables which are necessary in the biomechanical modeling approach, such as body joint angles, and weights of loads in hands, etc. However, the introduction of computer simulation has made it possible to

generate reasonable values of those variables from the PATH categorical values, and when integrated with biomechanical modeling, provide a better estimation of the physical requirements during construction work (Tak, 2005).

2.2. Biomechanical analysis technique

During the last 20 years, biomechanical models of the low back and shoulder have been developed and improved in an effort to obtain a more conclusive understanding of work-related musculoskeletal injury mechanisms (Davis and Jorgensen, 2005). Assuming that external moments imposed about the spine and shoulder joints are countered by the activities of trunk and shoulder musculatures, biomechanical models provide insight into the operation and behavior of trunk and shoulder muscle control systems. When accurate, those models can be used to both assess the loading imposed on the spine and shoulder joints during the performance of various tasks and predict the probability of injury or risk associated with a particular activity (Mirka and Marras, 1993).

In terms of the biomechanical analysis of manual material handling (MMH), one of the most common activities involved in drywall installation, there are three approaches generally used to solve for the muscle and joint forces required on the lumbar spine: 1) mathematical optimization, 2) electromyography (EMG) assisted, and 3) electromyography assisted by optimization which is a combination of the two previous approaches (Bean et al., 1988; Granata and Marras, 1995; Gagnon et al., 2001). Usually the basic trunk model includes 4 to 11 pairs of bilateral muscles, typically the lumbar erector spinae (LE), latissimus dorsi (LD), external oblique (EO), internal oblique (IO), and rectus abdominus (RA), and neglects other tissue contributions while the more

complicated models consist of 90 to 180 muscle fascicles with some of them accounting for ligament and passive tissue attributes. Since the unknown variables exceed the numbers of the force and moment equations in all cases, the solution for muscle and joint forces, and joint moments is indeterminate. With the three approaches listed above, the problems become solvable.

The optimization approach perfectly balances the net joint moment. One of the most important examples is the Double Linear Program developed by Bean et al. (1988). By minimizing the muscle contraction intensity and the joint compression force sequentially, the required muscle forces can be determined. The optimization approach always assumes there is no muscle co-contraction.

Contrarily, the EMG assisted approach predicts the muscle forces using an equation involving the normalized EMG data, a relative length parameter, and a relative velocity parameter, without perfectly balancing the net joint moment. The EMG assisted by optimization approach combines the advantages of the two methods, balancing perfectly the net joint moment while yielding the closest match between the predicted individual muscle forces and their mechanically optimal values (Gagnon et al., 2001).

There also exist specific instruments and techniques for analyzing manual material handling, especially its biomechanical influence on the lumbar spine. The Lumbar Motion Monitor (LMM) developed by Marras et al. (1995) allows for assessment of the instantaneous changes of the thoracolumbar spine in three-dimensional space. It is designed to be strapped on the back of a worker and to track the worker's trunk motion during work. The software that accompanies the LMM records information on trunk position, velocity, and acceleration, and also computes the probability of work-related

low back disorder high risk group membership by integrating data on liftrate (lifts/hour), average twisting velocity (deg/sec), maximum moment (Nm), maximum sagittal flexion (degrees), and maximum lateral velocity (deg/sec).

Ayoub et al. (1995) focused on the computerized human motion simulation using trajectory formation techniques. The predicted kinematic trajectories are based on generalization of the observed motion data from actual lifting activities. Muscle forces and moments are then calculated where the trajectories are used as inputs to the equations of motions. By minimizing the objective function of physical work, subject to the constraints, the trajectories can be fine-tuned to optimize the simulation.

The 3D Static Strength Prediction Program (3DSSPP) (University of Michigan, 1999) utilizes the Double Linear Program to predict the biomechanical and static strength capabilities of a specific person in relation to the physical demands of the work environment. Information such as the general anthropometry of the subject, the magnitude and direction of the load, whether one or two hands are involved, and the postural joint angles of the body relative to a horizontal reference axis system are inputs for the program, which estimates the required joint moment and strength, the strength capability percent, and the compressive force at the low back.

The shoulder models are relatively underdeveloped. In contrast to the lumbar spine, the complexity, mobility, and versatility of the shoulder complex has prohibited the development of guidelines for acceptable loads (Högfors et al., 1995). Some researchers, though, have tried to develop biomechanical models to predict the muscle forces and joint moments at the shoulder region (Högfors et al., 1987; Karlsson and Peterson, 1992; Van der Helm, 1994; Mohamed et al., 1996). Most models were used to

analyze load sharing between the muscles, the bones and the ligaments with different shoulder positions and different load situations. The equilibrium equations were derived from the balance of the external forces and the internal forces and their moments about the three main joints of the shoulder: the glenohumeral, the acromioclavicular and the sternoclavicular joints. Using the optimization technique and minimizing a selected objective function, the solution for the major muscles involved during the shoulder motion could be determined. Specifically, the muscles crossing the glehohumeral joint, such as the deltoideus, the infraspinatus, the supraspinatus, the subscapularis, the teres minor, the upper trapezius, and the biceps, were of more concern, because those muscles were most activated during shoulder abduction (Karlsson and Peterson, 1992).

2.3. Issues related to ergonomic exposure assessment

1) Exposure variability

It is noteworthy that many biomechanical models described above use single estimates (typically the sample mean) of input parameters, including muscle physiology (physiological cross-sectional area) and musculoskeletal geometry (moment arm), although it is generally said that these models solve the "general distribution problem" of portioning net joint moment between individual muscles (Hughes, 1997). There is great variability in anthropometric measures such as height and weight within the general population (Sanders and McCormick, 1987). Geometric and physiological parameters used in muscle force prediction models also vary widely within the population. Moreover, exposure levels (body postures and external loads) may vary between persons doing the same task and from one day to another for the same person (Punnett and Paquet, 1996;

Paquet et al., 2001). The differences in worker behavior or environmental parameters may also result in the variability in exposure. Thus, it is important to understand the distribution of these parameters and the effect of their variability on the predictions of muscle forces and joint forces and moments.

The Monte-Carlo simulation method is a technique whereby randomly generated numbers from a given distribution are used to simulate parameter variability (Semple et al., 2003). It can be applied to biomechanical models in order to specify the shape of the distribution for each parameter in the model and to set maximum and minimum values for each variable. Distributions are often specified as normal, lognormal, triangular, or uniform. The Monte-Carlo method has been successfully used both to capture the trunk muscle activity during torso bending (Mirka and Marras, 1993) and to simulate variability in muscle moment arms and physiological cross-sectional areas in the shoulder muscle force prediction (Hughes, 1997; Chang et al., 2000).

2) Load tolerance

While biomechanical models of the low back and shoulder provide estimates of the physical loads on the spinal structures and shoulder joints, the tolerance must also be known to understand the impact of the load magnitude. Disc compression is the most widely used tolerance limit for loads on the spine. It has been commonly accepted that compressive loads on the vertebral end plate of 3,400 N represent the level at which vertebral end plate micro-fractures begin to occur (Waters et al., 1993). Loadings greater than 6,400 N are considered to be hazardous to most workers (NIOSH, 1981). Factors such as gender, age, body density, disc degeneration, and number and frequency of

ergonomic exposures also influence the strength of the disc (Genaidy et al., 1993; Jager and Luttmann, 1992).

The failure tolerance of the intervertebral disc to anterior/posterior shear forces is estimated to range between 1,400 and 2,500 N (Yingling et al., 1997), whereas others have suggested the discs are at risk of sustaining a shear injury at loads as low as 1,000 N (McGill, 1997). These tolerances are also expected to be reduced with repetitive loading.

In terms of the load tolerance at the shoulder, Apreleva et al. (2000) documented that the largest reaction forces occurred when the ratio of applied forces favored the supraspinatus tendon, whereas simulated paralysis of the supraspinatus resulted in a significant decrease in joint compression. The magnitude of glenohumeral joint reaction forces varies according to the ratio of forces between the supraspinatus and deltoid muscles. Apreleva et al. (2000) measured reaction forces at the glenohumeral joint of approximately 335 N or 44% of total body weight at maximum abduction in the scapular plane. Inman et al. (1944) and Van der Helm (1994) predicted joint forces to be 50% of body weight and 330 N at 90° of abduction respectively, whereas Karlsson and Peterson (1992) predicted that the forces would reach a maximum of 650 N or 85% of body weight at 60° of abduction, according to a mathematical model of the full arm.

3) Muscle fatigue

Most biomechanical models of the low back and shoulder are applied in analysis of activities conducted in a short period, always assuming no fatigue or trying to minimize it. Nevertheless, when it is desired to investigate the physical ergonomic exposures during a relatively complex task which consists of different activities, it is

necessary to evaluate the cumulative effect of those activities. For example, it sounds plausible that the worker will feel weaker at the end of a workday, compared to the beginning of the day. It is also surmised that when the human body is challenged by some physical stress that does not cause outright acute failure of the system, numerous time-dependent biomechanical and physiological changes occur in response to the stress, namely "muscle fatigue" (Krajcarski, 2000).

On the other hand, the human body can recover during breaks or when conducting less strenuous activities. The challenge-recovery mechanism of muscle fatigue has thus provided a fundamental guideline for both quantification of muscle fatigue and development of optimal work-rest schedules (Von Rohmert, 1973; Fisher et al., 1993; Wood et al., 1997).

Significant research has been done to determine both which and how the trunk and shoulder muscles fatigue (O'Brien and Potvin, 1997; Sparto et al., 1997; Sparto and Parnianpour, 1998; Chen et al., 1999; Tsai et al., 2003). The lumbar erector spinae and the rotator cuff muscles are usually the focus when examining trunk and shoulder muscle fatigue respectively. Parameters used to describe fatigue including exertion level (percent of Maximum Voluntary Contraction), muscle decay rate, and muscle recovery rate, could be determined by the endurance-time based approach, or so-called "Rohmert curve" (Von Rohmert, 1960).

Most of the biomechanical analysis strategies addressed above are based on relatively simplified experiments or simulations in the laboratory. They usually involve simple tasks and fixed loads for all work cycles, without considering environmental factors. However, the construction industry is far more complicated than the lab. Its work

environment usually includes unique characteristics, such as constantly changing tasks/activities, materials, work site layout, and environmental conditions. Thus, it is difficult to apply objective ergonomic assessment instruments, such as EMG, LMM, and force platform; on the other hand, it requires that multiple methodologies be utilized together for more comprehensive evaluation. For example, during drywall installation, the installers perform different tasks/activities including cut/measure, MMH, and screw; the weight and size of drywall sheets differ according to the job requirements; while the work pace is affected by both the project schedule and personal expertise. Also, the environmental conditions may vary even between different work cycles, such as distance to walk, place to install, hindrance to walking or positioning, etc. Moreover, the cumulative effects on muscle force and joint resultant force requirements may be modified by factors such as muscle fatigue, whole-body balance, and spinal stability, etc.

3. Prior biomechanical analysis of drywall installation

Biomechanical analyses of drywall installation have been conducted previously by Pan and Chiou (1999) and Pan et al. (2002/2003). Using the 3DSSPP, Pan and Chiou (1999) analyzed the biomechanical stresses at the low back for the four most often observed drywall lifting techniques during installation. It was found that all four lifting techniques created overexertion hazards at the shoulders, torsos, and hips. Only a limited percentage of the male population had sufficient strength capability to lift a sheet weighing 45 kg. The method where the worker used one hand to support the horizontal drywall sheet at its bottom and the other hand to grasp the sheet at its top was the most stressful. The estimated L5/S1 and L4/L5 disc compression forces were consistently high

for all of the four lifting methods under various load situations. The results from this study indicated a necessity to identify less stressful drywall lifting methods and to develop safe assistive devices to reduce overexertion injuries.

Pan et al. (2002/2003) examined the effects of the four different drywall lifting methods on the installers' postural stability during a lab-based simulation. Both center-of-pressure (COP) and center-of-mass (COM) data were analyzed using univariate analyses and principle component analyses. Horizontal lifting with both hands on top of the drywall seemed to be most biomechanically efficient to reduce manual drywall handling hazards associated with fall potential and overexertion injuries. The authors also proposed work-practice intervention methods, such as job rotation, rest periods, team-work applications, and the use of assistive devices, as possible solutions to limit exposure to such hazards.

What remain to be investigated are both the ergonomic exposures when handling drywall and performing other activities during drywall installation, including cut/measure and screw using a powered tool, and the cumulative effects of the entire work process. The comparisons between the output values of biomechanical models and the load tolerance limits at the low back and shoulder are needed to explain the physical requirements for drywall installation. Quantifying muscle fatigue is also important to identify the potential adverse health effects from the ergonomic exposures. Further, the proposed intervention strategies need to be evaluated and compared against each other to prioritize them for field trials.

4. Significance of this research

Due to the dynamicity and complexity of construction work, observational ergonomic exposure assessment approaches tend to be prioritized to quantify ergonomic hazards. However, most of those approaches are time-consuming and labor-intensive, without offering sufficient quantification of all physical requirements of the construction work. By integrating the observational work-sampling based approach (PATH), Monte-Carlo simulation, and biomechanical models of the low back and shoulder, this research aimed to estimate the required muscle contraction forces and joint resultant forces during drywall installation. This approach should yield a better understanding of the physical hazards at the low back and shoulder for drywall installers.

The variability of the ergonomic exposures for different drywall installers was also examined by simulating different work-day activity series for different subjects. Based on the selection of different hypothesized anthropometric parameters, the physical loads on the low back and shoulder could be determined, respectively.

Considering the potential muscle fatigue during this tough construction work, this research defined and identified the prevalence of muscle fatigue, and used it as a proxy to delineate the physical requirement of drywall installation. Consecutive muscle fatigue periods were also computed to examine the concentrated facts of muscle fatigue.

This study provided a theoretical estimation of pre-and-post- interventions by comparing muscle forces, joint resultant forces, and muscle fatigue parameters. If such variables could be considered reasonable to represent physical ergonomic exposures during drywall installation, reduction and even minimization of those features might be used as guidelines for intervention evaluation and implementation.

C. Organization of Dissertation

Chapter II describes the estimates of required muscle contraction forces and joint resultant forces at the lumbar spine (L4/L5) and shoulder during a simulated 8-hour workday. It started with generating independent variables using the Monte-Carlo simulation method based on PATH data. Utilizing the model of the lumbar torso musculature developed by Marras and Granata (1995) and the "straight-line" shoulder muscle model developed by Högfors et al. (1987), muscle force vectors and moment arms were determined. Double linear optimization program and a nonlinear optimization program were then executed in MATLAB, from which the muscle forces and joint resultant forces at L4/L5 and shoulder joints of glenohumeral and sternoclavicular were obtained, respectively.

Chapter III examines the potential muscle fatigue at the low back and shoulder during the simulated workday. Muscle force production capacity was defined and compared to the required muscle contraction forces. The prevalences of muscle fatigue, the number and average duration of consecutive fatigue periods were computed to describe the physical requirements of the work.

Chapter IV evaluates the efficacies of proposed ergonomic interventions. The output values in the previous two chapters, including the required muscle contraction forces, joint resultant forces, prevalences of muscle fatigue, and consecutive fatigue periods, were selected as variables to be compared between the scenarios of pre- and post- intervention. Based on those comparisons, the biomechanical advantages and disadvantages of different intervention strategies were described.

The overall conclusions are presented in Chapter V, showing the integration of different methodologies to evaluate physical ergonomic exposures during drywall installation. Research limitations and recommendations for future work are also identified, with a focus on promoting drywall installers' occupational safety and health.

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II. Estimation of muscle contraction forces and joint resultant forces at the low back and shoulder during drywall installation

A. Introduction

A number of epidemiologic studies have examined the relationship between drywall installation and musculoskeletal injuries and disorders (Chiou et al., 2000; Lipscomb et al., 1997 and 2000). The low back is one of the most frequently injured body parts for drywall installers, especially during handling of the heavy drywall sheets.

The shoulder is another body part that is frequently injured for drywall installers (Lemasters et al., 1998; Lipscomb et al., 1997). Possible explanations include lifting of the heavy drywall sheets and driving screws overhead, especially when the distance between the external load and the body is far. Additionally, workers have to abduct the shoulder frequently during cutting and measuring drywall sheets and this might exacerbate the physical burden on the shoulder.

Significant efforts have been made to quantify the dose-response relationships between physical loads on the low back and adverse health effects including low back pain and back disorders. Particularly, the concept of dose or so-called "internal exposure" was introduced to address the physical work load in terms of biomechanical events inside the body (Winkel and Mathiassen, 1994). Biomechanical modeling became one means to assess the magnitude of physical stress to the musculoskeletal system, given that input values, such as anthropometric data, joint angles, external forces, and internal muscle parameters could be appropriately estimated.

Compared to the low back, biomechanical models of the shoulder are less developed because the shoulder is much more flexible and complex. It is also more

difficult to apply direct measurement instruments since there are a variety of bones, ligaments, muscles and other structures around the shoulder joints. In order to quantify the physical loads that are distributed on all anatomic parts of the shoulder complex, a simplified shoulder model is necessary to be able to predict muscle, ligament and joint resultant forces as functions of arm position and external loads in static or quasi-static situations (Engin, 1980; Högfors et al., 1987). When the effects of inertia and its distribution under motion and force are considered, a dynamic analysis of the shoulder mechanism becomes required (Van der Helm, 1994).

Direct measurement instruments, such as EMG (electromyography) and LMM (Lumbar Motion Monitor), may provide more objective and accurate assessments of the physical loads. Nevertheless, it is problematic to attach those instruments to construction workers during various tasks and/or for long time periods, simply because the construction industry is dynamic and complex. Thus, biomechanical modeling may be the optimal approach to describe the physical requirements during construction work.

With the development and application of the PATH methodology, an observational work-sampling based approach (Buchholz et al., 1996), it became practical to quantify the percent of time that the construction workers are exposed to awkward postures, various tasks and activities, and manual handling of different tools and materials (Punnett and Paquet, 1996; Paquet et al., 1996 and 2001). Although the categories of PATH, such as body part postures and loads in the hand, are not discrete, they have provided some basic information for use as input for biomechanical models of the low back and shoulder (Tak, 2005).

Considering the variability of different ergonomic exposures for different subjects over different tasks and times, many different input values are needed for the biomechanical models. The Monte-Carlo simulation method, a mathematic approach to random number generation based on assumed distributions, can be utilized to extract the categorical PATH data and generate discrete values for the biomechanical analysis.

The objective of this study was to integrate the methods of computer simulation, observational work sampling, and biomechanical modeling for a better understanding of the physical loads on the lumbar spine (L4/L5) and shoulder. The required muscle contraction forces and joint resultant forces at the low back and shoulder during a simulated 8-hour drywall installation work were estimated utilizing biomechanical models based on the Monte-Carlo simulation from PATH data.

B. Methods

1. Overview of methods

The study describes an integrating pathway through which the muscle contraction forces and joint resultant forces at the low back and shoulder during drywall installation were estimated (Figure 1). PATH, an observational work-sampling based approach, provided the fundamental delineation of drywall installation work by quantifying the percent of time that the drywall installers were conducting different activities with different body part postures. An 8-hour-workday activity series was constructed using Monte-Carlo simulation based on activity percentages from the PATH data (Step A in Figure 1). The biomechanical model input variables, including the anthropometric data, the joint angles, the external load force and position vectors, and the internal muscle parameters, were then generated for the analyses of the low back and shoulder (Steps B1 and B2 respectively in Figure 1). Utilizing different optimization programs in MATLAB, the three-dimensional static equilibrium equations were solved and the biomechanical model output variables of muscle contraction forces and joint resultant forces at the low back and shoulder were computed and summarized in the end (Steps C1 and C2 respectively in Figure 1).

2. Observational work sampling – PATH

A total of 126 PATH data points at 60-second intervals were collected by two well-trained researchers from COHP on a crew of 8 drywall workers from T. J. McCartney Inc. The researchers used PDAs (Casio E-200) to record the PATH observations. The PATH template (Appendix A) was programmed onto those PDAs

using InspectWrite (Penfact Inc.). The observations were made from Oct. 2003 to Jan. 2004 at the condominium construction site owned by Bovis Lend Lease in Park Plaza, Boston, MA, U.S.A.

For the purpose of the biomechanical analyses, seven kinds of activities were selected to represent the drywall installation task (Figure 2). Activity 1 (cut/measure) was defined as "Worker uses both cutter and tape measure to cut the whole drywall sheet into pieces that fit the structure." Activity 2 (lift), activity 3 (carry), and activity 4 (hold/place) denoted different components of manual material handling, and they were defined as "Worker moves the whole drywall sheet upwards without walking"; "Worker moves the drywall sheet with walking"; and "Worker stands still with the whole drywall sheet in the hands and makes adjustment of the sheet's position", respectively. Activity 5 (screw) meant that "Worker uses powered screwdriver to drive screws in to fasten the structure." Activity 6 (in between) stood for "Worker loads screw into the screwdriver and/or waits to begin driving screw." Particularly, activity 5 and 6 were assumed to always follow each other. Activity 7 (other) included climb/descend, communicate, scrape drywall sheets, mark/draw, etc., which were considered potentially insignificant and thus less interesting to analyze in terms of ergonomic hazards.

3. Generation of 8-hour-workday activity series

Two different kinds of work cycles of drywall installation were identified, based on field observations: 1) a whole sheet of drywall (Sheetrock® Brand Gypsum Panel from CGC Inc., assuming 1.22-m (4-ft) wide, 2.44-m (8-ft) long and 15.9-mm (5/8-in)

thick, with bulk density of 881 kg/m^3 (55 lb/ft^3)), and 2) a partial piece being installed with different methods.

For the whole sheet of drywall, the worker first lifts it from storage, where the drywall sheets are assumed to be piled horizontally on the ground (Step A in Figure 3). The worker then carries it to the hanging location with the postures shown in Step B (Figure 3). The third step is to rotate the drywall sheet and to place it at the destination (Step C in Figure 3). Last, the worker drives screws into the drywall sheet to fasten to the structure (Step D in Figure 3).

When the destination does not allow a whole piece of drywall sheet to be installed, the worker measures the structure first and cuts the whole piece to get a partial one. During drywall cutting, the drywall sheet is assumed to be stored vertically on the cart that is preliminarily conducted by the laborers based on the field observations (Step A in Figure 4). The worker then lifts and carries the partial piece of drywall sheet to the destination (Step B in Figure 4), and holds it in place while driving screws (Step C in Figure 4). Depending on the size of the partial drywall sheet, the numbers of screws required to fasten it will differ.

Based on the principle of the PATH methodology as a fixed-interval snapshot, the frequencies of different activities literally represent the probabilities of different activities. On the other hand, they may also serve as proxies for the relative durations of each activity, based on the assumption that the higher frequencies denote the relative longer durations. The structure of the activity series could be drawn according to the relative ratios of the chances of the two activities that possibly happen at a specific moment, assuming the average duration for every single activity (Figure 5).

From the PATH observations, it was determined that there are 12 possible work cycles, with 4 representing installation of a whole sheet and 8 denoting installation of a partial piece (Figure 5). The chance of starting a cycle for the installation of a whole sheet or a partial piece is the same, since the frequency of activity 3 (carry) was almost twice as that of activity 2 (lift) in PATH data (Figure 2) and activity 2 (lift) was rarely observed during the installation of partial sheets due to its short duration. The probability of each work cycle could be calculated by multiplying the probability of every single activity during that cycle. For example, the probability of work cycle of lift-carry-hold/place-screw-in-between (2-3-4-5-6) was 50%*100%*100%*45%*55% = 12%, where 45% meant the probability of conducting activity 5-6 (screw-in-between) at the specific moment when the subject could be possibly faced with the choices of two different kinds of activities, activity 5-6 or activity 7. Similarly, the probability of work cycle, cut/measure-carry-other-screw-in-between (1-3-7-5-6), was 50%*37%*55%*100%*55% = 6%.

The basic time unit was taken as 6 seconds (0.1 min), so for an 8-hour workday, 8*60*60/6 = 4800 time units need to be assigned to activities. Since different work cycles consisted of different activities and had different durations in reality, the activity code 0 was introduced to help maintain the same duration for each work cycle during the simulation, and it was deleted later in the output spreadsheet. It was learned from field interviews of drywall installers that one worker usually installs about 50 pieces of whole drywall sheets and some other partial sheets in a typical workday. Thus, one typical work cycle (for either a whole or partial sheet) was assumed to include 55 time units and 100

work cycles were simulated (55*100 = 5500 > 4800), considering that the activity code 0 needed to be deleted later.

The relative duration of each single activity included in the study was then assumed based on the ratios of PATH activity frequencies and the assumed total work cycles performed by the subject in the 8-hour workday. For example, when the average duration of activity 2 (lift) was defined as 5 time units, the average duration of activity 1 (cut/measure) was 3 to 4 times that of activity 2 (lift) since the frequencies of those two activities in PATH were 19.8% and 6.4%, respectively (Figure 2).

The 8-hour-workday activity series were simulated by the Monte-Carlo simulation method, assuming that the overall probabilities of the 12 possible courses of activity series followed the general discrete distribution.

4. Biomechanical analyses of the low back and shoulder

The biomechanical analyses of the low back and shoulder were conducted separately based on different requirements of the biomechanical models of the lumbar spine (L4/L5) and shoulder.

4.1. Biomechanical model of the lumbar spine (L4/L5)

4.1.1. Coordinate system

The coordinate system is aligned with the upper body of the subject. Its origin is located in the joint center of the lumbar spine at L4/L5. When the subject is standing erectly, the X-axis is defined laterally pointing from left to right, the Y-axis is directed forwards, and the Z-axis is directed upwards and perpendicular to X and Y.

When the trunk is moving, the coordinate system is rotated correspondingly following the Cardan angles in the order of Z, Y, and X. In that sense, the external load force and position vectors get transformed, and the internal muscle force and muscle-generated moment vectors keep constant instead. The joint resultant forces of disc compression, lateral shear, and anterior-posterior shear can be easily calculated based on the force equilibrium equation.

4.1.2. Generation of independent variables

1) L4/L5 Joint angles

Basically, L4/L5 joint angles are physically the same as the Cardan angles described above. The numeric values of L4/L5 joint angles were simulated from the categorical PATH data (Table 1). There are five categories in terms of trunk posture in PATH, which delineate the trunk motions in three dimensions. Specifically, i is defined as trunk flexion, which is the rotation about the X-axis; j is defined as lateral bend, which means the rotation about the Y-axis; whereas k is defined as twist, which represents the rotation about the Z-axis.

The rotation matrix of the coordinate system is thus determined as:

$$T = \begin{bmatrix} \cos j * \cos k & -\cos i * \sin k + \sin i * \sin j * \cos k & \sin i * \sin k + \cos i * \sin j * \cos k \\ \cos j * \sin k & \cos i * \cos k + \sin i * \sin j * \sin k & -\sin i * \cos k + \cos i * \sin j * \sin k \\ -\sin j & \sin i * \cos j & \cos i * \cos j \end{bmatrix} \tag{2.1}$$

Table 2 displays different ranges of the trunk posture categories of PATH, based on its definition. It is noteworthy that while trunk flexion follows one direction, trunk lateral bend and twist have twp possible directions. Unfortunately there are no data

describing how likely trunk lateral bend or trunk twist points to one direction versus another. However, it might be a reasonable assumption that for a right-handed person, the trunk tends to lean toward the left side and twist anticlockwise for most of time during the manual work. Given that the majority of the population is right-handed, an 80% of probability of such trunk lateral bend and twist motions was assumed.

Different distributions were assumed for each of the three trunk angles. A random number was then generated by the Monte-Carlo simulation method within each distribution.

A lognormal distribution was assumed for trunk flexion (Tak, 2005), since the values of trunk flexion angles were always positive according to the ranges for the PATH categories (Table 2). Its geometric mean and geometric standard deviation were calculated (Table 3) based on the rule of using the median value of every range for trunk flexion categories to represent such a range.

Normal distributions were assumed for trunk lateral bend and twist, and their mean and standard deviation were calculated respectively (Table 3) in the similar way as what was done for trunk flexion, except for the assumption of 80% probability of trunk bend toward the left side and twist anticlockwise.

2) External load position and force vectors

Different values of the external load positions on the X- and Y- axes were assumed when the subject was conducting different activities and using different trunk postures (Table 4). Since the whole sheets of drywall were assumed to lie horizontally on

the ground, the external load position for the Y-axis during the activity 2 (lift) was simply calculated as half of the sheet width.

Values of the external load position on the Z-axis were simulated in accordance with the PATH trunk flexion data. Within the range for each trunk flexion category (Table 2), the median value was selected. The mean and standard deviation were then calculated for the purpose of simulation and a normal distribution was assumed (Table 3).

In order to simulate the external load force vectors, especially their values on the Z-axis, PATH data on loads in hand were extracted (Table 5). A lognormal distribution was assumed. Similar method of simulating the trunk flexion angles was applied to determine the geometric mean and geometric standard deviation (Table 3). Since values for the weight of tools during activity 1, 5 and 6, and the weight of materials during activity 2 and 4 were known, random number generation was needed only for activity 3 and 7.

During activity 1 (cut/measure), there is also a "drag" force along the X-axis, which acts against the cutting process. It was assumed to be 10 N, 5 times the assumed weight of the cutter (Drywall Hammer Cutter), in the absence of information on the true value.

Moreover, during activity 5 (screw), both a horizontal reaction force along the Y-axis and a reaction torque about the Y-axis were added due to the manipulation of the drywall screwdriver (DeWalt Heavy-Duty VSR Drywall Screwdriver – DW252, weighing 12 N). Their values were assumed as 10 N and 7 N*m respectively, based on the screwdriver's specifications.

3) Anthropometric data

Summary statistics (Table 6) for subject weight, height, trunk widths and depths was acquired from Marras et al. (2001) (Subject anthropometry had not been obtained when the PATH data were collected). Subjects were all assumed to be healthy right-handed males with no history of either low back pain or chronic injuries. Assuming normal distributions for those anthropometric parameters and considering consistent correlations among each other, values of the 5, 50 (average), and 95 percentiles of those distributions were calculated to represent both the average and the extreme anthropometric characteristics (Table 7). The required muscle contraction forces and joint resultant forces for these three subjects thus represent the range of biomechanical loads likely to be experienced by U.S. construction workers.

In reality, the correlations between different anthropometric parameters may be quite variable; for example, people who are taller than others may not always be heavier. In order to examine the physical loads of drywall installation in general, 10 random values were simulated from those distributions without correlations and were assumed to fall in the specific ranges (Table 6). Those ranges were defined to represent the common anthropometries of construction workers.

The distance coefficient between the upper body center of mass (CM) and the L4/L5 joint based on subject height, and the percentages of the upper body weight of the whole body, were calculated respectively (Table 8), based on information from Drillis and Contini (1966) and Dempster (1955).

4.1.3. Biomechanical modeling

Based on previous EMG-assisted models, Marras and Granata (1995) developed a model of the lumbar torso musculature (Figure 6). Trunk kinetics was described by active muscle forces without considering passive muscle, ligament, and disc moments. This model was used to determine muscle force vectors and moment arms during different activities in the simulated 8-hour workday.

In this model, 10 muscle equivalent vectors, representing the left and right pairs of lumbar erector spinae (LEL and LER), latissimus dorsi (LDL and LDR), external oblique (EOL and EOR), internal oblique (IOL and IOR), and rectus abdominus (RAL and RAR), were determined from three-dimensional locations of muscle origins and insertions. Muscle forces are considered as vector quantities between their two endpoints

$$\vec{F}_m = |F_m| * \frac{\vec{r}_o - \vec{r}_i}{|\vec{r}_o - \vec{r}_i|} \tag{2.2}$$

where:
$\vec{F}_m$ = Muscle force vector
$|F_m|$ = Muscle force magnitude
$\vec{r}_o$ = Muscle origin vector
$\vec{r}_i$ = Muscle insertion vector

As shown in Figure 6, muscle origins denote the head of the force vector coplanar with the lumbar spine at L4/L5, whereas muscle insertions represent the tail of a force vector coplanar with the xyphoid process. The coefficients of muscle coordinates were displayed in Table 9. Vectors of both muscle origins and muscle insertions could be determined when values of subject trunk widths and depths at both muscle origin plane and muscle insertion plane were simulated.

$$\vec{r}_o = \begin{bmatrix} TW1 & 0 & 0 \\ 0 & TD1 & 0 \\ 0 & 0 & 1 \end{bmatrix} * \begin{bmatrix} X_o \\ Y_o \\ Z_o \end{bmatrix} \quad and \quad \vec{r}_i = \begin{bmatrix} TW2 & 0 & 0 \\ 0 & TD2 & 0 \\ 0 & 0 & 1 \end{bmatrix} * \begin{bmatrix} X_i \\ Y_i \\ Z_i \end{bmatrix} \tag{2.3}$$

where:
TW1 = Trunk width at muscle origins
TD1 = Trunk depth at muscle origins
TW2 = Trunk width at muscle insertions
TD2 = Trunk depth at muscle insertions
X_o, Y_o, Z_o = Coefficients of muscle coordinates at origins
X_i, Y_i, Z_i = Coefficients of muscle coordinates at insertions

Using linear regression methods, the relationship between the muscle Physiological Cross-Sectional Area (PCSA) and anthropometric parameters (Table 9), such as the subject height and weight, body mass index, and trunk width and depth at the planes of muscle origins and insertions, were also determined (Marras et al., 2001).

The external forces in the model include the weight of the upper body and the hand-held load. Joint resultant forces of disc compression, lateral shear, and anterior-posterior shear are calculated from the vector sum of muscle contraction forces and the external forces, based on the force equilibrium equation.

$$\sum \vec{F} = \vec{R} + \sum \vec{F}_m + \sum \vec{F}_{el} = 0 \tag{2.4}$$

where:
$\vec{R}$ = Joint resultant force vector in three dimensions, with each representing disc compression force, lateral shear force, and anterior-posterior shear force, respectively
$\sum \vec{F}_m$ = Sum of muscle contraction force vectors
$\sum \vec{F}_{el}$ = Sum of external force vectors

Muscle-generated moments about the spinal axis are predicted from the sum of vector products accumulating contraction forces of each muscle, j, and its moment arm.

$$\vec{M}_m = \sum_j \vec{r}_j * \vec{F}_j \tag{2.5}$$

where:
$\vec{M}_m$ = Muscle-generated moments
$\vec{r}_j$ = Moment arm vector of muscle j
$\vec{F}_j$ = Contraction force vector of muscle j

The moment equilibrium equation is set up by balancing muscle-generated moments and the external-load-generated moments.

$$\sum M = \vec{M}_m + \vec{M}_{el} = \vec{M}_m + \sum(\vec{r}_{el} * \vec{F}_{el}) = 0 \tag{2.6}$$

Where:
$\vec{M}_m$ = Muscle-generated moments
$\vec{M}_{el}$ = External-load-generated moments
$\vec{r}_{el}$ = External load position vector
$\vec{F}_{el}$ = External load force vector

There are 13 unknown variables, 10 muscle contraction forces and 3 joint resultant forces, which cannot be calculated by the six equilibrium equations. The Double Linear Program proposed originally by An et al. (1984) and then by Bean et al. (1988) was applied in this study to solve the indeterminate equation system.

The first linear optimization problem is to minimize the Maximum Muscle Contraction Intensity (MMCI) to resist the external moment.

$$\begin{aligned} &\min I \\ &subject\ to: \vec{M}_m + \vec{M}_{el} = \sum_{i=1}^{10} \vec{r}_i * \vec{F}_i + \vec{M}_{el} = 0 \\ &\qquad I \geq F_i / PCSA_i \\ &\qquad F_i \geq 0 \\ &\qquad for\ i = 1,2,...,10. \end{aligned} \tag{2.7}$$

The second objective function is to minimize the sum of muscle contraction forces, given a fixed maximum intensity obtained from the first optimization.

$$\begin{aligned} &\min \sum_{j=1}^{10} F_j \\ &subject\ to: \vec{M}_m + \vec{M}_{el} = \sum_{j=1}^{10} \vec{r}_j * \vec{F}_j + \vec{M}_{el} = 0 \\ &\qquad F_j \le I * PCSA_j \\ &\qquad F_j \ge 0 \\ &\qquad for\ j = 1,2,...,10 \end{aligned} \tag{2.8}$$

MATLAB (The MathWorks, Natick, MA, U.S.A.), a numeric computation package, was utilized to execute the Double Linear Program (Appendix B).

4.2. Biomechanical model of the shoulder

4.2.1. Coordinate system

One global coordinate system and four local coordinate systems were identified for the estimation of both muscle forces and joint resultant forces. The origin of the global coordinate system is coincident with the center of the glenohumeral (GH) joint of the right shoulder. Its X-axis points laterally from left to right. The Y- and Z- axes are directed forwards and upwards respectively.

In order to capture the relative motions of different shoulder elements, four local coordinate systems, one body (thorax) fixed system and three bone-fixed ones attached with clavicula, scapula, and humerus, are necessary (Högfors et al., 1987; Van der Helm, 1997; Wu et al., 2005) (Figure 7).

The thorax system has its origin in the center of the sternoclavicular (SC) joint. When the subject is standing erectly, the X_t-axis goes through the middle of both articular

surfaces and is directed to the right in the sagittal plane, and the Y_t- and Z_t- axes are directed forwards and upwards respectively. The thorax system moves along with the subject's trunk.

The clavicula system has the same origin as the thorax system. The X_c-axis goes through the center of the acromial articular surface. The Y_c-axis is orthogonal to the X_c-axis and parallel to the upper planar surface on the lateral end of the clavicula. The Z_c-axis is perpendicular to X_c and Y_c, pointing upwards.

The scapula system originates in the center of angulus acromialis (AA). The X_s-axis is directed through the inferior angle. The Y_s-axis is perpendicular to the X_s-axis and goes through the midpoint of the triangular surface on the medial border of the scapular in line with the scapular spine. The Z_s-axis is perpendicular to X_s and Y_s, pointing upwards.

The last coordinate system is fixed with the humerus, originating in the center of GH. Its X_h-axis is directed along the humerus through the end of the ridge between the coronoid fossa and the radial fossa. The Y_h-axis lies in the plane of the X_h-axis and the angular mobility direction of the ulna. The Z_h-axis is perpendicular to X_h and Y_h.

4.2.2. Generation of independent variables

1) Joint angles of shoulder, elbow and wrist

The external load positions change along with the body part movement. Coordinates of muscle and ligament origins and insertions around the shoulder joints also vary for different shoulder postures. Joint angles of trunk, shoulder, elbow and wrist need to be quantified during different activities. The trunk motion angles were simulated above.

Similarly for shoulder, elbow and wrist, angles representing their rotations about the three axes were defined as *i*, *j*, and *k* respectively. The external load positions for different activities with different postures could be calculated, using the position vector under neutral postures multiplying the following rotation matrix:

$$T_1 = \begin{bmatrix} \cos j * \cos k & \cos j * \sin k & -\sin j \\ -\cos i * \sin k + \sin i * \sin j * \cos k & \cos i * \cos k + \sin i * \sin j * \sin k & \sin i * \cos j \\ \sin i * \sin k + \cos i * \sin j * \cos k & -\sin i * \cos k + \cos i * \sin j * \sin k & \cos i * \cos j \end{bmatrix} \quad (2.9)$$

PATH has three categories to describe shoulder postures in terms of flexion: two arms down, 1 arm up, and 2 arms up. The critical value is a 60-degree angle between the arm and the trunk in the sagittal plane. Table 10 summarizes the frequencies of shoulder postures associated with the different activities. In order to generate a numeric value, specific ranges of *i*, *j*, and *k* for shoulder, elbow, and wrist were firstly assigned (Table 11). Different distributions were then assumed and the distribution parameters were determined (Table 12).

The distribution for shoulder flexion (angle *i*) was assumed to be lognormal. Using the median value of each range for *i-shoulder* to represent such a range, the geometric mean and geometric standard deviation could be calculated, assuming 72% of the arm use was assigned to the right arm when the arm posture was coded as "1 arm up" during PATH data collection (Porac and Coren, 1981; Tak, 2005).

The distributions for other motions of shoulder, elbow and wrist were all assumed as triangular ones, since there was only limited information about the motion ranges and the modal value within each range could be approximated. In detail, the modal value for *j-shoulder* was assumed to be 45 degrees, which meant the shoulder was most likely

abducted in the 45-degree plane corresponding to the coronal plane. During activity 5 (screw) and 6 (in between), *j-shoulder* equaled zero assuming no abduction.

The modal value for *i-elbow* was assumed to be 90 degrees and each single value was bigger than *i-shoulder*. The difference between those two was used to locate the positions of the forearm and hand in the sagittal plane. For *j-elbow*, the mode was 0, which meant that the forearm and the upper arm were most likely in the same line. During activity 5 (screw), *j-elbow* was assumed to be always zero.

The modal values for the remaining triangular distributions were all assumed to be 0. During activity 2 (lift) and 3 (carry), *k-wrist* was assumed -90 degrees because of a concern for holding the drywall sheets firmly. During activity 5 (screw), *j-wrist* was assumed 0 constantly in order to operate the screwdriver powerfully.

The coordinates of muscle and ligament origins and insertions around the shoulder joints were calculated slightly differently, where *j-shoulder* is the rotation about Z-axis representing the angle of abduction plane. Thus, i, j, and k compose a set of Euler angles. The rotation matrix is defined as:

$$T_2 = \begin{bmatrix} \cos k * \cos j - \sin k * \cos i * \sin j & \cos k * \sin j + \sin k * \cos i * \cos j & \sin k * \sin i \\ -\sin k * \cos j - \cos k * \cos i * \sin j & -\sin k * \sin j + \cos k * \cos i * \cos j & \cos k * \sin i \\ \sin i * \sin j & -\sin i * \cos j & \cos i \end{bmatrix} \quad (2.10)$$

2) Euler angles of coordinate system rotation

The interplay between the motions of the different parts of the shoulder is known as the "shoulder rhythm". Högfors et al. (1991) examined the shoulder rhythm by determining the ranges of Euler angles (α, β, γ) of the thorax system oriented along the

bone-fixed coordinate systems (Table 13). Since there is lack of information on the relationship between the value of the Euler angle and a specific shoulder motion, the Euler angles were simulated using the Monte-Carlo method. Using the median value to represent each range, the means and standard deviations of different Euler angles for different bones were calculated (Table 14). Random values could then be simulated, assuming the Euler angles follow normal distributions with those means and standard deviations and fall in the ranges illustrated in Table 13.

3) External load force vectors

Since subjects were all right-handed, the external load force vectors defined in the biomechanical analysis of the low back could be applied to the shoulder analysis for activities that were conducted by the right hand exclusively. During activity 2 (lift), 3 (carry), 4 (hold/place) and 7 (other) when both two hands were required to be in use, it was assumed that the external loads were equally distributed in two hands since the drywall sheets were symmetric.

4) Anthropometric data

Ten subjects were randomly simulated from the same distributions of subject height and weight as established for the low back analysis, assuming no correlations between those two parameters. All of the subjects were assumed healthy without history of shoulder injuries and illnesses. Since both the relative position relationship between different joints and the musculature-associated parameters (including origins, insertions, and PCSAs) were based on average values from previous research, it became

unnecessary to examine and compare the 5, 50 and 95 percentiles of those distributions with consistent correlation.

The weight percentages of different body parts of the whole body and the distance coefficients between the body part center of mass and the proximal joint are listed in Table 8.

4.2.3. Biomechanical modeling

An idealized mechanical model of the shoulder, presented by Högfors et al. (1987) and tested by Karlsson and Peterson (1992), was utilized in this study (Figure 8). In the model, the shoulder bones are considered as rigid bodies, and the shoulder joints are modeled as functional ball joints. The action of most of the muscles is described as one straight line following the shortest path between attachment points, except for the latissimus dorsi, the pectoralis major, the deltoideus, the infraspinatus, and the subscapularis, which require multiple lines. The joint resultant forces and ligament forces are the other two kinds of internal forces acting on the bones that are included in the model.

A total of 22 internal force variables were estimated: 19 major muscles attached to the humerus; the coracohumeral ligament, which is assumed to be able to produce an overall moment; and 2 resultant force vectors at GH and SC joints. Table 15 lists the coordinates of muscle and ligament origins and insertions in four local coordinate systems acquired from Högfors et al. (1987) and muscle PCSAs obtained from Karlsson and Peterson (1992).

In order to study the complete geometry of the shoulder, the relationship between the positions of the different joints needs to be determined (Klein-Breteler et al., 1999). The Euler angles (α, -β, γ) in the rotation order of Z-, Y-, and X- axes describing the "shoulder rhythm" were examined by Högfors et al. (1991). The transformation of the muscle and ligament coordinates into the global coordinate system could then be achieved by the inverse operation (-γ, β, -α) of orienting a bone-fixed system along the thorax system. The transformation matrix is written as:

$$R = \begin{bmatrix} \cos\beta * \cos\gamma & -\cos\beta * \sin\gamma & -\sin\beta \\ \cos\alpha * \sin\gamma - \sin\alpha * \sin\beta * \cos\gamma & \cos\alpha * \cos\gamma + \sin\alpha * \sin\beta * \sin\gamma & -\sin\alpha * \cos\beta \\ \sin\alpha * \sin\gamma + \cos\alpha * \sin\beta * \cos\gamma & \sin\alpha * \cos\gamma - \cos\alpha * \sin\beta * \sin\gamma & \cos\alpha * \cos\beta \end{bmatrix} \tag{2.11}$$

The muscle and ligament origin and insertion coordinates in the global coordinate system can then be calculated using the following equations.

$$\vec{r}_o = \vec{Q} + R * \vec{A}_o \quad and \quad \vec{r}_i = \vec{Q} + R * \vec{A}_i \tag{2.12}$$

Where:

$\vec{r}_o$ = Muscle origin coordinate in the global coordinate system

$\vec{r}_i$ = Muscle insertion coordinate in the global coordinate system

$\vec{Q}$ = The coordinate of the origin of the local coordinate system in the global coordinate system, which is determined by the relationship between the relative positions of different joints

$\vec{A}_o$ = Muscle origin coordinate in the local coordinate system

$\vec{A}_i$ = Muscle insertion coordinate in the local coordinate system

Similar to the biomechanical model of the lumbar spine, muscle and ligament forces could be determined using equation (2.2).

For analysis of the GH joint, the external forces include the weight of the upper arm, the weight of the lower arm, and the weight of the hand and the hand-held load,

whereas for the SC joint, the weights of the scapula and clavicula should be added. Force and moment equilibrium equations could thus be established as:

$$\begin{cases} \sum \vec{F} = \vec{R} + \sum \vec{F}_m + \sum \vec{F}_{el} = 0 \\ \sum M = \vec{M}_m + \vec{M}_{el} = \vec{M}_m + \sum (\vec{r}_{el} * \vec{F}_{el}) = 0 \end{cases} \tag{2.13}$$

where:

$\vec{R}$ = Joint resultant force vector with a magnitude $\sqrt{R_X^2 + R_Y^2 + R_Z^2}$

$\sum \vec{F}_m$ = Sum of muscle contraction force vectors

$\sum \vec{F}_{el}$ = Sum of external force vectors

$\vec{M}_m$ = Muscle-generated moments

$\vec{M}_{el}$ = External-load-generated moments

$\vec{r}_{el}$ = External load position vector

$\vec{F}_{el}$ = External load force vector

Optimization is required to solve the indeterminate problem. Based on previous research (Dul et al., 1984; Karlsson and Peterson, 1992; Högfors et al., 1995), the widely accepted objective function, which is to minimize the sum of the squared muscle stresses, was used in this study.

$$\begin{aligned} &\min \sum (F_i / PCSA_i)^2 \\ &subject\ to: \vec{M}_m + \vec{M}_{el} = \sum_{i=1}^{20} \vec{r}_i * \vec{F}_i + \vec{M}_{el} = 0 \\ &\qquad 0 \le F_i \le F_{i\max} \\ &\qquad F_{i\max} = k * PCSA_i \\ &\qquad for\ i = 1,2,...,20 \end{aligned} \tag{2.14}$$

Here k is a constant which depends on the maximum tension in muscles. Its value has been suggested to be in the range 40-100 N/cm^2 (Crowninshield and Brand, 1981). According to Wood et al. (1989), the average maximum muscle stress is about 100 psi (pounds per square inch), which is equivalent to 70 N/cm^2. Thus, k was assigned 70 N/cm^2 in this study. The ligament force was defined between zero and 10,000 N

(Mohamed et al., 1996). The MATLAB optimization toolbox was utilized to run such a nonlinear optimization program (Appendix C).

5. Data analysis

5.1. Low back

Anthropometric data on body height and weight, trunk widths and depths at both muscle origins and muscle insertions for 10 subjects that were randomly simulated from assumed distributions were analyzed. Histograms of simulated 8-hour-workday activities for the selected 5, 50, and 95 percentiles of the hypothesized population and the 10 randomly generated subjects were drawn and compared to the frequencies of PATH activity data.

Summary statistics of the MMCI (Maximum Muscle Contraction Intensity), the calculated muscle contraction forces and joint resultant forces for the three different percentiles of the hypothesized population and the average of the 10 randomly generated subjects were obtained. Disc compression forces were compared against the load tolerance values proposed by NIOSH (1981) and Waters et al. (1993), where frequencies of disc compression forces above the load tolerance values for the simulated 8-hour workday were calculated.

5.2. Shoulder

Anthropometric data of subject height and weight for the 10 randomly generated subjects were analyzed, and the simulated 8-hour-workday activities for the 10 subjects were compared to the frequencies of observed PATH data. Statistical information about the required muscle and ligament contraction forces and resultant forces at the

glenohumeral (GH) and sternoclavicular (SC) joints was summarized, including the mean, median, standard deviation or geometric standard deviation, minimum, and maximum.

C. Results

1. Low back analysis

1.1. Description of 10 simulated subjects

The 10 hypothetical drywall installers averaged 176.2 (SD 9.8) cm in height and 83.1 (SD 10.6) kg in weight (Table 16). Figure 9 and Figure 10 illustrate the histograms for simulated 8-hour-workday activities for the selected 5, 50, and 95 percentiles of the hypothesized population and the 10 randomly generated subjects, respectively. In general, the simulations generated similar values to the frequencies of observed PATH activities.

1.2. Maximum muscle contraction intensity (MMCI)

The means of MMCI for the 5, 50, and 95 percentiles of the hypothesized population were 70.5, 43.6, and 35.9 N/cm^2, respectively (Table 17 and Figure 11). It can be clearly seen from Figure 11 that while the minimum values of MMCI were the same for all three subjects based on the definition of the Double Linear Program, the other statistical parameters, including the mean, standard error, median, standard deviation, range, and maximum value, followed a decreasing trend by percentile. Particularly, the maximum value of MMCI for the 5 percentile of the hypothesized population was 367.8 N/cm^2, which was impossibly high and needs further explanation.

The average of MMCI means for 10 simulated subjects was 41.0 N/cm^2, which fell in the range from 35.9 to 70.5 N/cm^2 for the hypothesized population (Table 18 and Figure 12). The average of MMCI standard deviations was 30.6 N/cm^2, which was understandable since subjects conducted various activities requiring different muscle

strengths during the 8-hour workday. Notably, the average of MMCI maximum values was 178.2 N/cm^2 and the maximum reached 251.9 N/cm^2.

1.3. Required muscle contraction forces

For the 5, 50, and 95 percentiles of the hypothesized population, both the mean and maximum values of muscle forces for LEL, LER, LDL, and LDR followed a decreasing trend (Table 19 and Figures 13-16). Once again, the maximum muscle contraction forces of LEL and LER for the 5 percentile were 7119.9 and 7363.6 N respectively, which were greater than the published maximum value estimated by Farfan (1973).

The averages of LEL, LER, LDL, and LDR contraction force means for 10 simulated subjects during the simulated 8-hour workday were 989.7, 1009.7, 591.0, and 603.8 N, respectively. The averages of the maximum contraction forces for those four muscle groups were 4785.9, 4686.4, 3363.4, and 3727.8 N, respectively (Table 20 and Figures 17-19).

1.4. Joint resultant forces

The means of disc compression forces for the 5, 50, and 95 percentiles of the hypothesized population were 1606.4, 1729.0, and 1925.0 N, respectively (Table 21 and Figures 20-23). They followed an increasing trend by percentile, whereas the maximum values of disc compression forces followed a decreasing trend, as did the absolute maximum values of lateral shear forces and anterior-posterior shear forces. The means and medians of the absolute values of lateral shear forces and anterior-posterior shear

forces for the three percentiles were similar. When compared to the load tolerance values of 3400 N and 6400 N recommended by NIOSH, the prevalences of calculated disc compression forces above the tolerances during the 8-hour workday were 10.7% and 2.5% for the 5 percentile, 8.5% and 0.7% for the 50 percentile, and 8.4% and 0.4% for the 95 percentile, respectively, indicating decreasing trends (Figure 24).

Table 22 and Figures 25-27 show the statistical results of joint resultant forces for the 10 simulated subjects. The averages of means of disc compression forces, lateral shear forces and anterior-posterior shear forces (absolute values) were 1724.9, 291.1, and 487.5 N, respectively. The averages of geometric standard deviations of those three joint resultant forces were 0.57, 1.08, and 1.28 N respectively with a small standard deviation. In addition, the averages of maximum values of joint resultant forces were 6899.9, 2385.7, and 3038.0 N respectively, and the highest value for disc compression force was found to be 7784.8 N.

There was 8.5% of time during the 8-hour workday, on average, that the calculated disc compression forces exceeded the NIOSH action limit of 3400 N. When compared to the maximum permissible limit of 6400 N, the percentage went down to 0.8% (Figure 24).

2. Shoulder analysis

2.1. Description of 10 simulated subjects

The mean weight and height for the 10 randomly generated subjects were 81.6 (SD 6.5) kg and 175.5 (SD 5.8) cm. The simulated 8-hour-workday activities for the 10 subjects on average produced underestimates for activities 1 (cut/measure), 2 (lift), 3

(carry) and 7 (other), and overestimates for activities 5 (screw) and 6 (in between), compared to the PATH observations (Figure 28).

2.2. Required muscle contraction forces

. The coracohumeral ligament produced the highest values of force on average, in terms of both the mean (233.6 N) and the maximum (3257.5 N) (Table 23 and Figures 29-31). Muscles with high average force values included biceps brachii (short head), supraspinatus, teres major, and triceps, etc. The average maximum values of muscle forces generated by the three different parts of deltoideus were also notable. The average standard deviations for all muscle groups were fairly high, indicating big differences in the physical loads among different activities.

2.3. Joint resultant forces

The averages of mean joint resultant forces for the 10 simulated subjects were 480.9 (GSD 1.02) N and 574.4 (GSD 0.98) N for GH and SC, respectively (Table 24 and Figures 32-34). The averages of maximum values of joint resultant forces reached 4158.3 N and 4237.5 N, with the highest one approaching 5000 N.

D. Discussion

1. Key findings

The physical loads on the L4/L5 level of the lumbar spine and shoulder during drywall installation were estimated in this study by integrating the methods of PATH, Monte-Carlo simulation, and biomechanical modeling. The required muscle contractions forces and joint resultant forces at L4/L5 and the glenohumeral joint and sternoclavicular joint during a simulated 8-hour workday were calculated to delineate the physical ergonomic exposures for drywall installers. Those results have helped to explain why the low back and shoulder are the most frequently injured body parts for the workers.

For the sake of both ergonomic hazard identification and injury prevention, the data analysis approach in this project allowed an approximate estimation of the physical requirements at the low back and shoulder during one day of work. Although there were many assumptions involved that might have limited the validity of the results, the study did save labor and time costs tremendously compared to what would have been needed for direct measurements. It has also provided a baseline database for evaluating the efficacy of ergonomic interventions in the future.

2. Validity of simulation results

In general, the frequencies of simulated 8-hour-workday activities for the biomechanical analyses of the low back and shoulder were similar to the observed PATH data (Figures 9, 10 and 28). Since it was assumed that activities 5 (screw) and 6 (in between) always followed each other, their frequencies during the 8-hour workday simulation were almost equal. The overestimates of those two activities potentially

caused the underestimates of activity 7 (other). Similar patterns applied to activities 2 (lift) and 4 (hold/place), because the subject would have to hold/place the whole drywall sheet after he lifted it from the ground and carried it to the destination. This might have also caused the underestimates of activity 3 (carry), since it was assumed that the probability of activity 3 (carry) was twice as that of activity 2 (lift) based on PATH data. Similarly under that scenario, the chance whether to start a work cycle of a whole sheet installation or a partial one was assumed to be equal, when ignoring activity 2 (lift) in the partial sheet installation cycles because of its little occurrence during PATH observations. Nevertheless, it should be recognized that in reality, the installation of partial drywall sheets may involve all kinds of MMH activities that were studied in the whole sheet installation, including lift, carry, and hold/place. Compared to the installation of the whole drywall sheets, there was less concern of ergonomic hazards when conducting activities of lift and hold/place during the installation of those partial sheets. Thus, only activity of carry was selected to represent the MMH activities during those work cycles. This might have underestimated the overall physical requirements of MMH activities posed on the drywall installers.

The basic time unit was assumed to be 6 seconds (0.1 min) during the simulation, which equaled one tenth of the PATH data collection interval (1 min). Such a selection was aimed at tracking the subject motions elaborately, since it could be imagined that many different things could possibly happen within one minute. On the other hand, the length of basic time unit equaled to the time interval that was commonly selected in the one-compartmental model to study muscle fatigue. It should be noted that not all activities followed the same pace, since lifting the heavy and bulky drywall sheet from

the ground was usually observed faster than carrying and holding/placing. However, the average durations of those three activities were assumed to be the same based on the relative ratios of their frequencies in PATH. Given that the longer a certain activity lasted, the higher the probability that it might have been captured during PATH observations, the linkage between the frequencies of PATH activities and their relative durations were established. While the average durations of activity 2 (lift) might have been overestimated in the study, the use of short time interval and average durations were specifically helpful for the study of activity 5 (screw), because a number of work interface situations were involved which required different body part postures.

It should be noted that since there were only about 2 hours of PATH observations made, their representation of the actual drywall installation work might be problematic. There might be many more kinds of work cycles in reality besides the selected 12 ones. Fortunately activities selected to represent drywall installation were all quite physical and simple to simulate; however, the effects of different possible assumptions on the results still remain unknown and warrant further analysis, especially for activities that are complex to capture.

In this study, the subjects' anthropometric data were simulated based on Marras et al. (2001), since they were not collected during PATH observations. Based on the defined ranges for subject height (160 ~ 200 cm) and weight (65 ~ 110 kg), the randomly generated data were assumed to represent the "real" construction workers. Nonetheless, the anthropometries of construction workers may be fairly diverse in fact and thus may not be easily simulated.

3. Low back results

The calculated maximum values of MMCI (Tables 17 and 18) were much higher than the range from 35 to 137 N/cm^2 that has been usually reported (Buchanan, 1995). Buchanan (1995) concluded that the maximum muscle stress (similar as MMCI) was not a constant, based on the differences in specific tension in elbow flexors and extensors. It is surmised that the maximum stress for trunk muscles, especially the lumbar spinae, can be varied too.

Akurathi et al. (2002) found that the intensity of lumbar erector spinae for the knee level lift of a 10-kg load was about 100 N/cm^2. Considering that the whole drywall sheet selected in the study weighed more than 40 kg and was stored on the ground, it is not unreasonable to imagine higher MMCIs. Also, the combination of trunk postures during drywall lifting from the simulation in this study might have been too awkward to result in such high MMCI values. Nonetheless, the maximum value of MMCI for the 5 percentile of the hypothesized population exceeded 350 N/cm^2, which still seemed unrealistic. Given that the Double Linear Program does not set up the upper limit for MMCI, it requires more information to explain such an extreme result.

The study found that the simulated 8-hour drywall installation work required more physical effort for the lighter and shorter person compared to the heavier and higher one, particularly for conducting the manual handling activities which caused higher muscle contraction forces and joint resultant forces. Generally, a shorter person may bring the load being lifted closer to the body to be more protective. However, it is not the same scenario in this study since the distance between the whole drywall sheet and the body had to be one-half of the drywall sheet width, assuming that it was piled

horizontally on the ground. The shorter person had to bend his trunk more to reach the further end of the drywall sheet, which potentially posed a higher risk of low back injury for him.

4. Shoulder results

As expected and examined by previous researchers, there were a variety of muscles around the shoulder joints involved in the shoulder movement during installation of drywall. The coracohumeral ligament had the highest calculated force in general, which may have meant it was an important contributor to generate force and moment, if its assumed upper limit of 10,000 N is reasonable. Muscle groups with fairly high forces included biceps (short head), supraspinatus, teres major, triceps, and different parts of the deltoideus, etc., which are in accordance with the results from other investigations (Karlsson and Peterson, 1992; Siemienski et al., 1995).

It should be mentioned that several activities resulting in different types of shoulder motions were examined in this study, whereas in the compared studies, one simplified activity was usually selected. Similarly under that scenario, the Euler angles used to describe the "shoulder rhythm" might have been quite variable, instead of simply simulated based on the previous data from Högfors et al. (1991) in this study. Högfors et al. (1987) also defined constraints to some muscles during modeling the paths of muscle strings. However, those constraints were not considered in this study for the sake of simplified quantification of the physical ergonomic hazards. The results might have been more accurate, if those constraints had been included.

In addition, the biceps and triceps are usually responsible for the lower arm movement and are neither actually attached to the humerus. Since the influence of the lower arm on the shoulder complex had to be considered, the biceps and triceps were regarded to have lower attachments to the humerus depending on the elbow angle (Högfors et al., 1987).

The resultant force at the sternoclavicular joint was higher generally than the glenohumeral joint resultant force, which contradicts the result of Siemienski et al. (1995). One possible reason is that multiple activities besides screwdriving were analyzed in this study. Moreover, the application of muscles around the glenohumeral joint to the sternoclavicular joint and the exclusion of muscles around the sternoclavicular joint might be other explanations.

No acceptable levels of shoulder load tolerance have been publicized, simply because of the versatility and complexity of the shoulder structures. Many researchers, however, have tried to use the body weight (BW) as a criterion to determine the acceptableness of joint resultant forces, especially the glenohumeral joint. For example, Poppen and Walker (1978) suggested a value of 0.9 times BW for unloaded shoulder abduction based on a two-dimensional biomechanical model, whereas the maximum glenohumeral joint resultant force was calculated to be 0.85 times BW from simple abduction with a 1 kg load in Karlsson and Peterson (1992). More recently, Anglin et al. (2000) found that the average glenohumeral joint resultant forces ranged from 1.3 to 2.4 times BW during performance of the routine activities of daily living by older people.

Since this study utilized a method similar to that of Karlsson and Peterson (1992), body weight was selected as a critical value for the comparison of the calculated joint

resultant forces. It was found that the average frequencies of joint resultant forces above subject body weights were 16.8% (SD 1.3%) and 18.4% (SD 1.3%) respectively, for the glenohumeral joint and sternoclavicular joint. Such data could be considered during the evaluations of ergonomic intervention in the future.

5. The integration of different methods

The present study extracted information from PATH data and applied it to the simulation of every single subject. It was noteworthy that PATH data were collected on a crew of 8 workers and they thus represented the average structure of the work, which probably did not mean that everyone had the same frequencies of activities, trunk and shoulder postures, and loads in the hands. It might have been an ideal scenario in this study that everyone installed drywall in a similar pattern, although the actual values of activity percentages, trunk and shoulder joint angles, and weights handled varied.

In order to quantify the muscle contraction forces and joint resultant forces easily, a three-dimensional free-body diagram was applied to simplify the activities assuming there was no inertia. Analyses of force and moment were done only on the upper body. Since leg postures in the PATH data were neutral for a majority of time, they were neglected.

Trunk lateral bend and twist were not differentiated during collection of PATH data, because it was hard to distinguish those two different trunk postures in different planes at the same time. The simulation bundled the two together and thus might have overestimated the awkward trunk postures during the work.

In this study, the median value of every category of trunk postures and loads in the hands was used to represent such a category. The simulations were based on either a normal distribution or a lognormal one. Given that there may be other ways to select a representative value and assume a possible distribution, it would be interesting to explore and compare the sensitivity of different strategies in a future investigation.

Similar to the trunk posture, the shoulder posture categories of PATH method only provide information on shoulder flexion in the sagittal plane. More information is required for joint angle simulations of the shoulder, elbow and wrist in three dimensions. While triangular distributions for the majority of joint angles were assumed, the validity and sensitivity of such data analysis strategies need to be further studied.

Compared to the biomechanical analysis of the low back, the shoulder analysis did not find large inter-subject variability. We surmise that this was because we used average values of muscle parameters including coordinates of origins and insertions and PCSAs. It could be assumed that, on the other hand, subjects' anthropometric data, especially the muscle parameters, do contribute to the variability of physical loads on the shoulder.

We did not examine which independent variables better predict the required muscle forces and joint resultant forces. However, it would be worthy to study for the purpose of ergonomic interventions. For example, distances between the external load and the body were either assumed or simulated from assumed distributions. If these variables have greater influences on the results than other independent variables, especially during the activities lift and carry, interventions can be targeted on them potentially.

The advantages of biomechanical analysis approach in this study over 3DSSPP may be presented by taking the variability of the muscle parameters into consideration. While 3DSSPP assumes the same muscle moment arm and PCSA for different anthropometries, those variables were simulated based on the relationship with the anthropometric data (Marras et al., 2001). Thus, it might have provided more accurate prediction in terms of this concern.

E. Conclusions

The integration of ergonomic exposure assessment techniques, including observational work-sampling, computer simulation, and biomechanical modeling provided a feasible estimation of the physical loads on the low back and shoulder during drywall installation. The biomechanical model output values of the MMCI (Maximum Muscle Contraction Intensity), the required muscle and ligament forces, joint resultant forces at the lumbar spine (L4/L5) and the glenohumeral joint and sternoclavicular joint, and their comparisons against the suggested load tolerances, were determined to describe a simulated 8-hour workday and provide a database for further study of ergonomic intervention evaluations.

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Table 1

Frequencies of PATH trunk posture data associated with different activities

Activity	Amount	Neutral	Mild Flexion	Severe Flexion	Bend & Twist	Bend & Twist & Flexion	Activity Percent	Neutral	Mild Flexion	Severe Flexion	Bend & Twist	Bend & Twist & Flexion
1 (cut/measure)	25	12	4	3	1	5	19.8%	9.5%	3.2%	2.4%	0.8%	4.0%
2 (lift)	8	0	2	1	0	5	6.4%	0.0%	1.6%	0.8%	0.0%	4.0%
3 (carry)	16	15	0	1	0	0	12.7%	11.9%	0.0%	0.8%	0.0%	0.0%
4 (hold/place)	6	2	1	2	1	0	4.8%	1.6%	0.8%	1.6%	0.8%	0.0%
5 (screw)	20	7	5	3	3	2	15.9%	5.6%	4.0%	2.4%	2.4%	1.6%
6 (in between)	24	22	2	0	0	0	19.1%	17.5%	1.6%	0.0%	0.0%	0.0%
7 (other)	27	13	2	8	0	4	21.4%	10.3%	1.6%	6.4%	0.0%	3.2%
Total	126	71	16	18	5	16	100.0%	56.4%	12.7%	14.3%	4.0%	12.7%

Table 2

Ranges of PATH trunk posture categories and external load position vectors on the Z-axis

Category		i (degree)	j (degree)	k (degree)	ez (m)
Neutral		0 ~ 20	-20 ~ 20	-20 ~ 20	0.0 ~ 0.8
Mild Flexion		20 ~ 45	-20 ~ 20	-20 ~ 20	-0.4 ~ 0.2
Severe Flexion		45 ~ 80	-20 ~ 20	-20 ~ 20	-0.9 ~ -0.1
Bend & Twist	Left*	0 ~ 20	20 ~ 35	-45 ~ -20	0.0 ~ 0.8
	Right	0 ~ 20	-35 ~ -20	20 ~ 45	0.0 ~ 0.8
Bend & Twist & Flexion	Left	20 ~ 80	20 ~ 35	-45 ~ -20	-0.9 ~ 0.2
	Right	20 ~ 80	-35 ~ -20	20 ~ 45	-0.9 ~ 0.2

Note: i – Flexion; j – Lateral bend; k – Twist; ez – External load position vectors at Z-axis.

* – Assuming the trunk bended left and twisted anticlockwise (view from above) with 80% probability.

Table 3

Distribution parameters for trunk postures, external load position vectors on the Z-axis, and external load force vectors on the Z-axis

Variable		Distribution Type	Distribution Parameter	
			Mean/GM (degree)	SD/GSD (degree)
Trunk Posture	i	Lognormal	2.9	0.8
	j	Normal	2.8	10.9
	k	Normal	-3.2	12.9
Variable		**Distribution Type**	**Distribution Parameter**	
			Mean (m)	**SD (m)**
ez		Normal	0.11	0.37
Variable		**Distribution Type**	**Distribution Parameter**	
			GM (N)	**GSD (N)**
wz		Lognormal	3.0	1.2

Note: i – Flexion; j – Lateral bend; k – Twist; GM – Geometric Mean; SD – Standard Deviation; GSD – Geometric Standard Deviation; ez – External load position vectors on the Z-axis; wz – External load force vectors on the Z-axis.

Table 4

External load position vectors on the X- and Y- axes associated with different activities and trunk postures

Activity	(ex, ey) (m)				
	Trunk Posture				
	Neutral	Mild Flexion	Severe Flexion	Bend & Twist	Bend & Twist & Flexion
1 (cut/measure)	0.3, 0.3	0.3, 0.4	0.3, 0.5	0.3, 0.3	0.3, 0.4
2 (lift)	0.0, 0.6	0.0, 0.6	0.0, 0.6	0.0, 0.6	0.0, 0.6
3 (carry)	0.0, 0.3	0.0, 0.4	0.0, 0.5	0.0, 0.3	0.0, 0.4
4 (hold/place)	0.0, 0.2	0.0, 0.3	0.0, 0.4	0.0, 0.2	0.0, 0.3
5 (screw)	0.2, 0.3	0.2, 0.4	0.2, 0.5	0.2, 0.3	0.2, 0.4
6 (in between)	0.0, 0.3	0.0, 0.4	0.0, 0.5	0.0, 0.3	0.0, 0.4
7 (other)	0.0, 0.3	0.0, 0.4	0.0, 0.5	0.0, 0.3	0.0, 0.4

Note: ex – External load position vectors on the X-axis; ey – External load position vectors on the Y-axis.

Table 5

Frequencies and ranges of PATH loads in the hands data

Activity	Amount	Very Light (0~23)	Light (23~67)	Medium (67~223)	Heavy (223~408)	Activity Percent	Very Light (0~23)	Light (23~67)	Medium (67~223)	Heavy (223~408)
1 (cut/measure)	25	25	0	0	0	19.8%	19.8%	0.0%	0.0%	0.0%
2 (lift)	8	0	0	0	8	6.4%	0.0%	0.0%	0.0%	6.4%
3 (carry)	16	1	5	4	6	12.7%	0.8%	4.0%	3.2%	4.8%
4 (hold/place)	6	0	0	0	6	4.8%	0.0%	0.0%	0.0%	4.8%
5 (screw)	20	0	20	0	0	15.9%	0.0%	15.9%	0.0%	0.0%
6 (in between)	24	0	24	0	0	19.1%	0.0%	19.1%	0.0%	0.0%
7 (other)	27	27	0	0	0	21.4%	21.4%	0.0%	0.0%	0.0%
Total	126	53	49	4	20	100.0%	42.1%	38.9%	3.2%	15.9%

Table 6

Distribution parameters of selected subjects' anthropometric data*

Variable	Distribution Type	Distribution Parameter		Specific Range
		Mean	SD	
W	Normal	79.8 (kg)	13.3 (kg)	65 ~ 110 (kg)
H	Normal	175.9 (cm)	9.1 (cm)	160 ~ 200 (cm)
TW1	Normal	0.303 (m)	0.022 (m)	N/D
TD1	Normal	0.223 (m)	0.022 (m)	N/D
TW2	Normal	0.324 (m)	0.020 (m)	N/D
TD2	Normal	0.229 (m)	0.022 (m)	N/D

Note:

* – Based on Marras et al. (2001);

W – Weight;

H – Height;

TW1 – Trunk width coplanar with the lumbar spine at L4/L5 (muscle origins);

TD1 – Trunk depth coplanar with the lumbar spine at L4/L5 (muscle origins);

TW2 – Trunk width coplanar with the xyphoid process (muscle insertions);

TD2 – Trunk depth coplanar with the xyphoid process (muscle insertions);

N/D – Not defined.

Table 7

Anthropometric data of 5, 50, and 95 percentiles of the hypothesized population included in the biomechanical analysis of the low back

Category	Variable	Percentile		
		5	50	95
Anthropometry	W (kg)	57.9	79.8	101.7
	H (cm)	160.9	175.9	190.9
	TW1 (m)	0.267	0.303	0.339
	TD1 (m)	0.187	0.223	0.259
	TW2 (m)	0.291	0.324	0.357
	TD2 (m)	0.193	0.229	0.265
Trunk Muscle PCSA (cm^2)	LEL	19.36	26.45	30.49
	LER	20.02	26.37	29.98
	LDL	13.32	19.09	25.62
	LDR	16.70	21.71	26.73
	EOL	8.18	10.70	13.56
	EOR	8.50	10.85	13.50
	IOL	7.57	10.60	13.15
	IOR	7.73	10.27	12.81
	RAL	5.35	9.30	11.54
	RAR	5.50	9.29	11.45

Note: PCSA – Physiological Cross-Sectional Area; LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; LDL – Left Latissimus Dorsi; LDR – Right Latissimus Dorsi; EOL – Left External Oblique; EOR – Right External Oblique; IOL – Left Internal Oblique; IOR – Right Internal Oblique; RAL – Left Rectus Abdominus; RAR – Right Rectus Abdominus.

Table 8

Percentages of body part weights of the whole body and distance coefficients between body part centers of mass (CM) and the joints*

Body Part	Percentage of Whole Body Weight	Link	Distance Coefficient
Upper Body	47.6%	Upper-Body-CM – L4/L5	0.186
Upper Arm	2.8%	Upper Arm	0.186
Forearm	1.6%	Shoulder – Upper-Arm-CM	0.074
Hand	0.5%	Forearm	0.146
Clavicula	0.2%	Elbow – Forearm-CM	0.066
Scapula	0.9%	Hand	0.108
		Wrist – Hand-CM	0.040

Note: * – Based on Drillis and Contini (1966) and Dempster (1955).

Table 9

Coefficients of trunk muscle coordinates and PCSAs*

Muscle	PCSA (cm^2)	Origins			Insertions		
		X	Y	Z	X	Y	Z
LEL	53.65-12.34*H/W	-0.20	-0.30	0	-0.30	-0.30	Z**
LER	50.7-11.04*H/W	0.20	-0.30	0	0.30	-0.30	Z**
LDL	-4.65+0.032*TW2*TD2*10000	-0.25	-0.30	0	-0.60	0.10	Z**
LDR	3.44+0.229*W	0.25	-0.30	0	0.60	0.10	Z**
EOL	0.315+0.014*TW2*TD2*10000	-0.10	0.55	0	-0.45	-0.19	Z**
EOR	1.2+0.013*TW2*TD2*10000	0.10	0.55	0	0.45	-0.19	Z**
IOL	-4.05+32.29*W/H	-0.45	0.30	0	-0.45	0.20	Z**
IOR	1.01+0.116*W	0.45	0.30	0	0.45	0.20	Z**
RAL	24.44-6.87*H/W	-0.10	0.55	0	-0.10	0.55	Z**
RAR	23.84-6.6*H/W	0.10	0.55	0	0.10	0.55	Z**

Note:

* – Based on Marras and Granata (1995) and Marras et al. (2001);

** – Z** = -0.0275*H/100+0.30*TD2;

W – Weight; H – Height;

TW1 – Trunk width coplanar with the lumbar spine at L4/L5 (muscle origins);

TD1 – Trunk depth coplanar with the lumbar spine at L4/L5 (muscle origins);

TW2 – Trunk width coplanar with the xyphoid process (muscle insertions);

TD2 – Trunk depth coplanar with the xyphoid process (muscle insertions);

PCSA – Physiological Cross-Sectional Area; LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; LDL – Left Latissimus Dorsi; LDR – Right Latissimus Dorsi; EOL – Left External Oblique; EOR – Right External Oblique; IOL – Left Internal Oblique; IOR – Right Internal Oblique; RAL – Left Rectus Abdominus; RAR – Right Rectus Abdominus.

Table 10

Frequencies of PATH shoulder posture data associated with different activities

Activity	Amount	2 down	1 up	2 up	Percent	2 down	1 up	2 up
1 (cut/measure)	25	13	4	8	19.8%	10.3%	3.2%	6.4%
2 (lift)	8	4	1	3	6.4%	3.2%	0.8%	2.4%
3 (carry)	16	10	3	3	12.7%	7.9%	2.4%	2.4%
4 (hold/place)	6	3	0	3	4.8%	2.4%	0.0%	2.4%
5 (screw)	20	9	6	5	15.9%	7.1%	4.8%	4.0%
6 (in between)	24	23	0	1	19.1%	18.3%	0.0%	0.8%
7 (other)	27	18	2	7	21.4%	14.3%	1.6%	5.6%
Total	126	80	16	30	100.0%	63.5%	12.7%	23.8%

Table 11

Ranges of PATH shoulder postures and

assumed ranges of elbow and wrist postures

Joint	Category		i (degree)	j (degree)	k (degree)
Shoulder	Two arms down		0 ~ 60	0 ~ 90	-20 ~ 20
	One arm up	Left*	0 ~ 60	0 ~ 90	-20 ~ 20
		Right	60 ~ 150	0 ~ 90	-20 ~ 20
	Two arms up		60 ~ 150	0 ~ 90	-20 ~ 20
Elbow	Possible ranges		0 ~ 135	-90 ~ 30	-20 ~ 20
Wrist	Possible ranges		-20 ~ 20	-20 ~ 20	-20 ~ 20

Note: * – Assuming 72% of the arm use was assigned to right arm when the arm posture was coded as "One arm up".

Table 12

Distribution parameters of shoulder, elbow, and wrist postures

Variable	Distribution Type	Distribution Parameter	
		GM (degree)	GSD (degree)
i-shoulder	Lognormal	3.8	0.6

Variable	Distribution Type	Distribution Parameter		
		Min (degree)	Mode (degree)	Max (degree)
j-shoulder	Triangular	0	45	90
k-shoulder	Triangular	-20	0	20
i-elbow	Triangular	0	90	135
j-elbow	Triangular	-90	0	30
k-elbow	Triangular	-20	0	20
i-wrist	Triangular	-20	0	20
j-wrist	Triangular	-20	0	20
k-wrist	Triangular	-20	0	20

Note: i – Flexion; j – Abduction; k – Rotation.

Table 13

Ranges of Euler angles of bone-fixed coordinate system rotation for three subjects during the motion studies by Högfors et al. (1991)

Subject No.	Bone	α (degree)	β (degree)	γ (degree)
1	Clavicula	316.4 ~ 331.8	17.6 ~ 37.0	0.1 ~ 15.5
	Scapula	202.0 ~ 253.3	-70.7 ~ 88.0	78.1 ~ 122.9
	Humerus	19.4 ~ 62.5	-16.5 ~ 23.6	41.3 ~ 71.2
2	Clavicula	318.3 ~ 348.2	10.0 ~ 21.2	5.5 ~ 27.5
	Scapula	100.8 ~ 219.9	-74.7 ~ 88.2	0.0 ~ 115.7
	Humerus	-2.3 ~ 80.9	-34.6 ~ 8.4	68.6 ~ 88.0
3	Clavicula	300.0 ~ 323.0	15.5 ~ 47.2	-8.6 ~ 11.2
	Scapula	169.4 ~ 205.6	-61.6 ~ 82.2	71.7 ~ 103.3
	Humerus	8.7 ~ 59.6	-17.1 ~ 21.1	86.5 ~ 107.2

Table 14

Means and standard deviations of the Euler angles

Bone	α (degree)		β (degree)		γ (degree)	
	Mean	STD	Mean	STD	Mean	STD
Clavicula	323.0	10.9	24.8	8.2	8.5	7.6
Scapula	191.8	33.9	8.6	1.8	82.0	21.9
Humerus	38.1	3.5	-2.5	9.2	77.1	20.3

Table 15

Shoulder muscle coordinates in four local coordinate systems and PCSAs*

Muscle	PCSA (cm^2)	CS	Origins			CS	Insertions		
			X	Y	Z		X	Y	Z
LD (Superior)	2.61	t	0.260	0.000	-0.750	h	0.170	0.049	0.017
LD (Inferior)	3.15	t	0.298	-0.266	-0.809	h	0.170	0.049	0.017
PMa (Sternum)	4.50	t	0.000	0.148	-0.209	h	0.275	0.040	0.043
PMa (Clavicle)	2.75	c	0.374	0.097	0.027	h	0.275	0.040	0.043
DE (Anterior)	4.29	s	-0.009	-0.147	-0.062	h	0.369	-0.004	0.044
DE (Medial)	5.63	s	0.241	0.199	-0.174	h	0.369	-0.004	0.044
DE (Posterior)	5.21	c	0.795	-0.028	0.015	h	0.369	-0.004	0.044
CO	1.91	s	0.118	-0.014	0.193	h	0.396	0.014	-0.018
IN (Superior)	3.87	s	0.617	0.240	-0.106	h	-0.018	-0.025	0.094
IN (Inferior)	4.05	s	0.760	0.115	-0.069	h	-0.018	-0.025	0.094
Sub (Superior)	3.13	s	0.523	0.307	-0.137	h	0.012	0.085	0.011
Sub (Middle)	2.82	s	0.668	0.164	-0.108	h	0.012	0.085	0.011
Sub (Inferior)	3.87	s	0.762	0.069	-0.064	h	0.012	0.085	0.011
Sup	3.83	s	0.305	0.348	-0.112	h	-0.042	0.031	0.079
TMa	4.48	s	0.844	-0.058	-0.006	h	0.143	0.037	-0.022
TMi	1.89	s	0.600	-0.049	-0.024	h	0.039	-0.045	0.037
BB (Long)	2.14	s	0.102	0.016	0.014	h	1.036+0.143*cosρ	0.143*sinρ	0.000
BB (Short)	1.89	s	0.051	0.007	0.227	h	1.036+0.143*cosρ	0.143*sinρ	0.000
TRI	8.87	s	0.366	-0.109	0.041	h	1.036–0.64*sin(0.73–ρ)	-0.64*cos(0.73–ρ)	0.000
COL	N/A	s	0.080	0.080	0.260	h	-0.035	0.050	0.053

Note: * – Based on Högfors et al. (1987) and Karlsson and Peterson (1992); LD – Latissimus Dorsi; PMa – Pectoralis Major; DE – Deltoideus; CO – Coracobrachialis; IN – Infraspinatus; Sub – Subscapularis; Sup – Supraspinatus; TMa – Teres Major; TMi – Teres Minor; BB – Biceps Brachii; TRI – Triceps; COL – Coracohumeral Ligament; PCSA – Physiological Cross-Sectional Area; CS – Coordinate System; t – Thorax; c – Clavicula; s – Scapula; h – Humerus; ρ – The complement of the elbow angle in radians; N/A – Not Applicable.

Table 16

Statistical information about anthropometric parameters for the 10 subjects included in the biomechanical analysis of the low back

Parameter	H (cm)	W (kg)	TW1 (m)	TD1 (m)	TW2 (m)	TD2 (m)
Mean	176.2	83.1	0.306	0.232	0.317	0.236
Median	173.3	79.1	0.308	0.235	0.315	0.235
SD	9.8	10.6	0.013	0.015	0.011	0.017
Range	29.1	32.0	0.048	0.042	0.033	0.062
Min	163.9	65.7	0.282	0.207	0.303	0.203
Max	193.0	97.7	0.329	0.249	0.336	0.265

Note:

W – Weight;

H – Height;

TW1 – Trunk width coplanar with the lumbar spine at L4/L5 (muscle origins);

TD1 – Trunk depth coplanar with the lumbar spine at L4/L5 (muscle origins);

TW2 – Trunk width coplanar with the xyphoid process (muscle insertions);

TD2 – Trunk depth coplanar with the xyphoid process (muscle insertions).

Table 17

Statistical information about the MMCI (Maximum Muscle Contraction Intensity) for 5, 50, and 95 percentiles of the hypothesized population

Parameter	5%	50%	95%
Mean	70.5	43.6	35.9
SE	1.1	0.5	0.3
Median	41.0	28.8	24.9
SD	73.0	34.1	22.1
Range	345.7	172.7	115.0
Min	22.1	22.1	22.1
Max	367.8	194.8	137.1

Table 18

Statistical information about the MMCI (Maximum Muscle Contraction Intensity) for the 10 simulated subjects

Parameter	Mean	Median	SD	Max
Mean	41.0	27.5	30.6	178.2
SE	2.4	1.7	2.7	13.5
SD	7.7	5.4	8.5	42.8
Min	31.1	22.1	18.2	113.9
Max	54.2	36.5	45.4	251.9

Table 19

Statistical information about the calculated trunk muscle contraction forces for 5, 50, and 95 percentiles of the hypothesized population

Subject	Parameter	LEL	LER	LDL	LDR	EOL	EOR	IOL	IOR	RAL	RAR
5%	Mean	1331.9	1386.6	865.3	854.4	0.5	22.4	113.7	90.3	0.0	0.0
	SE	20.8	21.4	14.5	14.1	0.1	1.5	3.7	2.0	0.0	0.0
	Median	774.3	814.8	494.6	556.4	0.0	0.0	0.0	12.8	0.0	0.0
	SD	1435.1	1479.2	1002.1	974.5	9.0	106.6	253.4	135.9	0.6	1.0
	Max	7119.9	7363.6	4891.6	5511.3	260.9	1654.5	2293.6	1884.2	26.3	39.7
50%	Mean	1074.5	1082.8	648.1	665.2	0.5	5.1	91.2	93.6	0.0	0.2
	SE	14.0	13.8	10.6	10.8	0.1	0.6	2.7	2.0	0.0	0.1
	Median	727.4	751.5	478.1	529.6	0.0	0.0	0.0	17.5	0.0	0.0
	SD	958.4	949.1	727.7	743.6	8.8	39.7	184.4	137.3	1.0	3.7
	Max	5152.3	5135.9	3713.2	4048.6	249.4	767.0	1711.4	1403.0	51.8	143.9
95%	Mean	984.7	984.0	619.5	650.9	1.9	1.6	88.6	103.2	0.0	0.1
	SE	11.0	10.7	10.1	10.3	0.3	0.3	2.4	2.1	0.0	0.0
	Median	716.0	734.4	500.4	590.9	0.0	0.0	0.0	25.1	0.0	0.0
	SD	757.1	735.5	695.8	710.0	23.3	22.0	166.0	143.4	0.4	3.3
	Max	4179.7	4109.8	3500.8	3558.5	648.6	666.3	1541.4	1326.2	23.8	161.0

Note: LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; LDL – Left Latissimus Dorsi; LDR – Right Latissimus Dorsi; EOL – Left External Oblique; EOR – Right External Oblique; IOL – Left Internal Oblique; IOR – Right Internal Oblique; RAL – Left Rectus Abdominus; RAR – Right Rectus Abdominus.

Table 20

Statistical information about the calculated trunk muscle contraction forces for the 10 simulated subjects

Parameter		LEL	LER	LDL	LDR	EOL	EOR	IOL	IOR	RAL	RAR
Mean	Mean	989.7	1009.7	591.0	603.8	3.6	12.1	93.5	90.7	0.1	0.2
	SE	77.0	75.0	46.5	51.4	2.1	4.1	8.2	8.4	0.0	0.0
	SD	243.5	237.0	146.9	162.6	6.7	12.9	25.8	26.5	0.1	0.2
	Min	650.6	677.5	395.1	386.0	0.3	0.3	49.8	45.6	0.0	0.0
	Max	1359.7	1379.6	815.3	862.5	22.2	38.2	138.6	132.1	0.3	0.5
Median	Mean	699.9	729.1	419.8	441.9	0.0	0.0	0.0	11.5	0.0	0.0
	SE	49.2	48.9	46.1	59.5	0.0	0.0	0.0	3.8	0.0	0.0
	SD	155.6	154.7	145.8	188.1	0.0	0.0	0.0	12.0	0.0	0.0
	Min	485.5	516.0	148.1	122.6	0.0	0.0	0.0	0.0	0.0	0.0
	Max	934.2	949.8	604.8	683.0	0.0	0.0	0.0	29.1	0.0	0.0
SD	Mean	898.2	876.8	657.3	677.2	28.2	52.3	197.2	135.6	2.3	3.4
	SE	60.8	60.7	32.2	33.9	10.7	11.8	13.3	9.3	0.6	0.6
	SD	192.4	191.9	101.9	107.3	33.9	37.2	42.1	29.6	1.9	1.8
	Min	640.7	622.2	465.0	522.7	7.8	9.7	130.3	86.7	0.0	1.5
	Max	1261.0	1238.2	765.9	838.2	122.0	122.8	281.3	183.7	5.0	7.1
Max	Mean	4785.9	4686.4	3363.4	3727.8	522.2	969.0	1829.1	1344.8	91.4	140.9
	SE	346.9	322.9	182.8	256.0	95.5	125.4	134.5	63.5	21.9	17.6
	SD	1097.1	1021.1	578.0	809.7	302.0	396.6	425.5	200.7	69.4	55.6
	Min	3279.4	3255.5	2205.7	2536.9	240.8	484.3	1314.4	1049.7	0.0	72.1
	Max	6688.7	6413.3	3903.1	4883.4	1256.9	1590.1	2684.0	1592.8	194.5	214.0

Note: LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; LDL – Left Latissimus Dorsi; LDR – Right Latissimus Dorsi; EOL – Left External Oblique; EOR – Right External Oblique; IOL – Left Internal Oblique; IOR – Right Internal Oblique; RAL – Left Rectus Abdominus; RAR – Right Rectus Abdominus.

Table 21

Statistical information about the calculated joint resultant forces at L4/L5 for 5, 50, and 95 percentiles of the hypothesized population

Subject	Parameter	COMP	ABS(LS)	ABS(APS)
5%	Mean	1606.4	386.8	571.8
	Median	1058.8	299.4	290.9
	GSD	0.66	1.06	1.25
	Range	7116.6	3273.3	3919.6
	Min	314.0	0.2	0.0
	Max	7430.6	3273.5	3919.6
50%	Mean	1729.0	319.6	475.7
	Median	1312.8	259.6	271.6
	GSD	0.57	1.11	1.01
	Range	6438.4	2734.5	2865.4
	Min	491.2	0.0	1.2
	Max	6929.6	2734.5	2866.6
95%	Mean	1925.0	307.7	503.6
	Median	1573.3	254.1	293.9
	GSD	0.50	1.12	0.96
	Range	6217.6	2301.4	2787.2
	Min	611.6	0.0	0.3
	Max	6829.2	2301.4	2787.5

Note: COMP – Disc compression force; ABS(LS) – Absolute value of lateral shear force; ABS(APS) – Absolute value of anterior-posterior shear force.

Table 22

Statistical information about the calculated joint resultant forces at L4/L5 for the 10 simulated subjects

	Mean			Median		
	COMP	**ABS(LS)**	**ABS(APS)**	**COMP**	**ABS(LS)**	**ABS(APS)**
Mean	1724.9	291.1	487.5	1312.7	239.9	309.4
SE	64.9	23.9	93.9	65.1	19.7	60.6
SD	205.1	75.6	296.8	205.9	62.2	191.5
Min	1454.3	190.0	181.6	1048.3	150.7	136.6
Max	2050.8	423.9	1126.2	1628.6	346.8	726.3
	GSD			**Range**		
	COMP	**ABS(LS)**	**ABS(APS)**	**COMP**	**ABS(LS)**	**ABS(APS)**
Mean	0.57	1.08	1.28	6421.9	2385.7	3037.8
SE	0.01	0.02	0.08	159.3	223.6	540.7
SD	0.03	0.06	0.26	503.7	707.2	1710.0
Min	0.51	0.97	0.93	5656.3	1472.8	1169.4
Max	0.60	1.15	1.75	7292.4	3937.8	6636.2
	Min			**Max**		
	COMP	**ABS(LS)**	**ABS(APS)**	**COMP**	**ABS(LS)**	**ABS(APS)**
Mean	478.0	0.1	0.1	6899.9	2385.7	3038.0
SE	18.4	0.0	0.0	171.4	223.7	540.7
SD	58.1	0.1	0.1	542.0	707.3	1710.0
Min	397.9	0.0	0.0	6054.2	1472.8	1169.6
Max	586.4	0.2	0.5	7784.8	3938.0	6636.2

Note: COMP – Disc compression force; ABS(LS) – Absolute value of lateral shear force; ABS(APS) – Absolute value of anterior-posterior shear force.

Table 23

Statistical information about the calculated shoulder muscle contraction forces for the 10 simulated subjects

Parameter		LD(S)	LD(I)	PMa(S)	PMa(C)	DE(A)	DE(M)	DE(P)	CO	IN(S)	IN(I)
Mean	Mean	6.9	6.5	27.4	10.8	1.0	1.7	6.1	4.0	0.3	0.8
	SE	0.2	0.2	0.7	0.3	0.1	0.1	0.2	0.2	0.0	0.1
	SD	0.7	0.7	2.1	0.9	0.2	0.4	0.6	0.5	0.1	0.2
	Min	5.9	5.5	24.4	9.3	0.7	1.3	4.8	3.4	0.2	0.6
	Max	8.1	7.7	31.0	12.2	1.4	2.5	6.8	4.8	0.5	1.2
Median	Mean	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
	SE	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
	SD	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
	Min	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
	Max	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
SD	Mean	22.5	22.3	75.0	32.0	12.7	19.8	29.2	14.0	5.2	11.7
	SE	0.6	0.7	1.0	0.6	0.7	0.9	0.7	0.5	0.5	0.5
	SD	2.0	2.1	3.2	2.0	2.1	2.7	2.3	1.5	1.5	1.5
	Min	19.7	19.5	70.0	28.8	8.8	15.8	23.9	12.2	2.5	9.3
	Max	25.5	25.3	80.4	34.5	16.2	24.5	31.9	16.0	7.1	14.4
Max	Mean	207.6	245.4	341.1	233.8	351.9	428.6	432.3	197.2	234.5	330.8
	SE	3.4	3.4	3.4	6.0	7.9	5.3	10.0	8.2	30.3	7.6
	SD	10.7	10.7	10.6	19.0	25.0	16.8	31.6	26.0	95.7	24.1
	Min	189.3	227.1	321.6	213.4	300.5	402.9	364.9	154.6	70.0	288.0
	Max	221.6	259.4	353.9	265.7	385.9	450.7	476.7	245.7	332.4	369.1

Table 23 (cont.)

Statistical information about the calculated shoulder muscle contraction forces for the 10 simulated subjects

Parameter		Sub(S)	Sub(M)	Sub(I)	Sup	TMa	TMi	BB(L)	BB(S)	TRI	COL
Mean	Mean	25.2	23.6	30.5	61.5	52.4	6.0	16.8	77.2	51.5	233.6
	SE	0.4	0.3	0.5	0.7	0.7	0.1	0.3	0.8	0.7	4.3
	SD	1.3	1.1	1.6	2.2	2.1	0.3	0.9	2.7	2.1	13.6
	Min	23.7	21.7	28.4	59.0	49.5	5.4	15.5	73.5	47.1	213.7
	Max	27.4	24.7	32.7	66.5	55.0	6.4	18.1	81.9	53.9	249.7
Median	Mean	1.2	1.3	2.3	45.4	9.0	0.1	0.5	73.8	16.2	64.6
	SE	0.0	0.1	0.1	0.7	0.1	0.0	0.0	1.3	0.4	1.2
	SD	0.2	0.2	0.3	2.2	0.5	0.0	0.1	4.0	1.2	3.6
	Min	0.8	0.8	1.6	43.3	8.1	0.1	0.4	68.3	14.5	60.5
	Max	1.3	1.5	2.6	50.0	9.7	0.1	0.6	80.9	18.5	72.5
SD	Mean	56.1	51.5	69.0	68.1	100.3	21.9	41.4	42.7	101.1	466.6
	SE	0.6	0.5	0.7	0.4	0.7	0.3	0.4	0.1	1.4	7.9
	SD	1.9	1.5	2.3	1.3	2.1	0.8	1.2	0.4	4.5	24.9
	Min	54.1	49.4	66.2	65.4	97.6	20.1	39.6	41.8	91.8	427.7
	Max	59.2	53.1	72.5	70.0	103.8	23.1	43.1	43.2	106.1	500.4
Max	Mean	294.3	273.8	347.2	289.7	387.2	191.7	226.2	209.2	666.8	3257.5
	SE	6.7	6.4	6.4	3.6	7.4	7.4	6.3	6.6	8.0	123.3
	SD	21.2	20.3	20.4	11.5	23.4	23.3	20.0	21.0	25.2	390.0
	Min	257.5	235.8	309.3	271.9	345.5	139.8	188.5	170.7	621.9	2786.9
	Max	331.1	309.4	382.9	306.2	425.6	217.9	261.8	244.3	702.9	3928.4

Note: LD(S) – Latissimus Dorsi (Superior); LD(I) – Latissimus Dorsi (Inferior); PMa(S) – Pectoralis Major (Sternum); PMa(C) – Pectoralis Major (Clavicle); *DE(A) – Deltoideus (Anterior); DE(M) – Deltoideus (Medial); DE(P) – Deltoideus (Posterior); CO – Coracobrachialis;* IN(S) – Infraspinatus (Superior); IN(I) – Infraspinatus (Inferior); Sub(S) – Subscapularis (Superior); Sub(M) – Subscapularis (Middle); Sub(I) – Subscapularis (Inferior); Sup – Supraspinatus; TMa – Teres Major; TMi – Teres Minor; BB(L) – Biceps Brachii (Long); BB(S) – Biceps Brachii (Short); TRI – Triceps; COL – Coracohumeral Ligament.

Table 24

Statistical information about the calculated joint resultant forces at GH (glenohumeral) and SC (sternoclavicular) joints for the 10 simulated subjects

	Mean		Median	
	FGH	**FSC**	**FGH**	**FSC**
Mean	480.9	574.4	219.0	270.1
SE	7.0	7.7	3.0	3.6
SD	22.1	24.5	9.3	11.4
Min	443.8	534.9	204.4	256.2
Max	504.8	603.3	235.5	293.7
	GSD		**Range**	
	FGH	**FSC**	**FGH**	**FSC**
Mean	1.02	0.89	4138.7	4182.0
SE	0.01	0.01	164.7	139.1
SD	0.02	0.02	520.9	439.7
Min	0.99	0.85	3391.0	3502.3
Max	1.05	0.93	4978.9	4864.1
	Min		**Max**	
	FGH	**FSC**	**FGH**	**FSC**
Mean	19.6	55.5	4158.3	4237.5
SE	1.9	1.9	164.0	138.8
SD	5.9	5.9	518.6	438.9
Min	11.3	45.3	3407.7	3557.7
Max	30.5	67.0	4993.1	4916.0

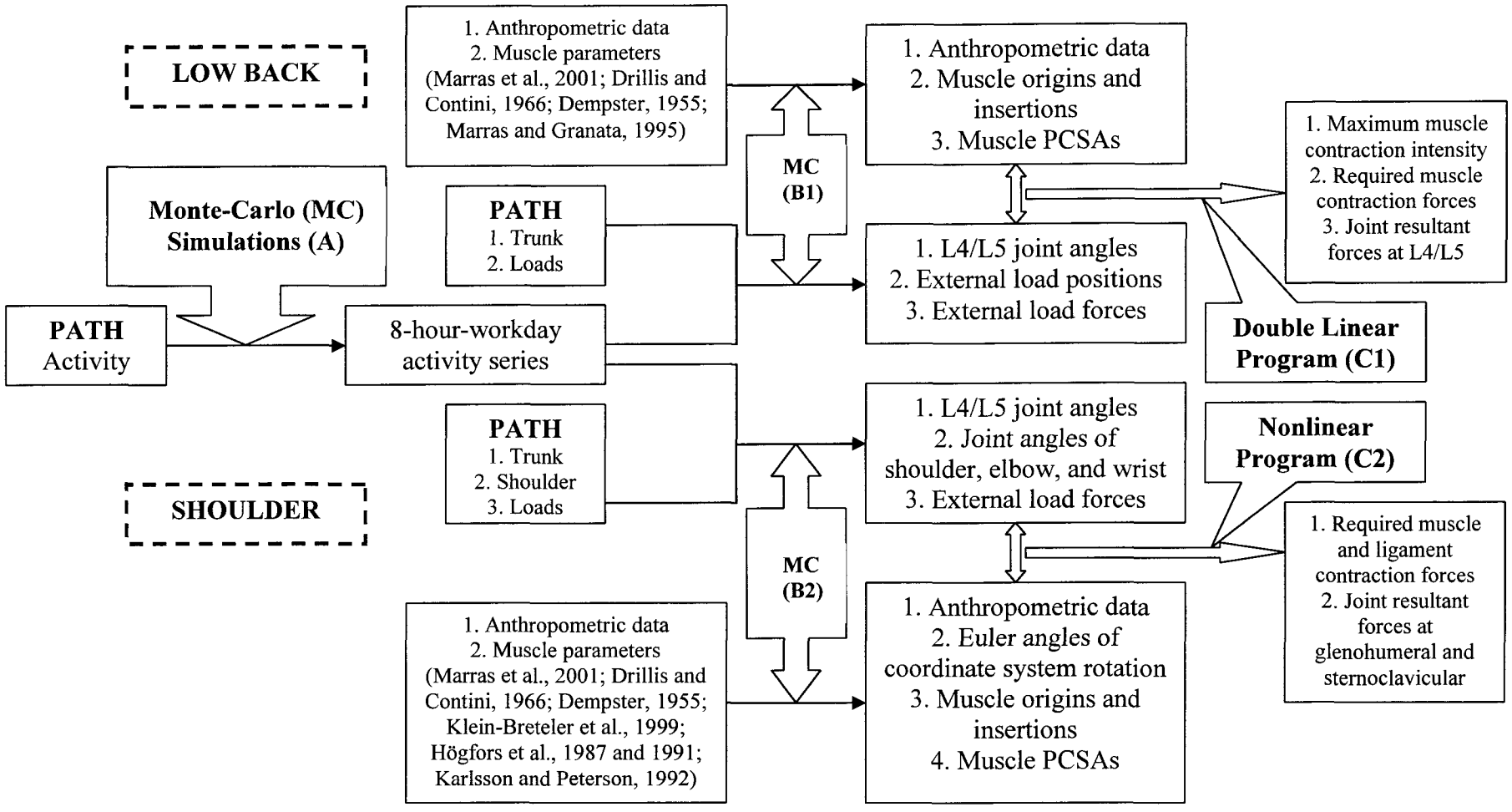


Figure 1 Schematic diagram for data analysis procedure in the study

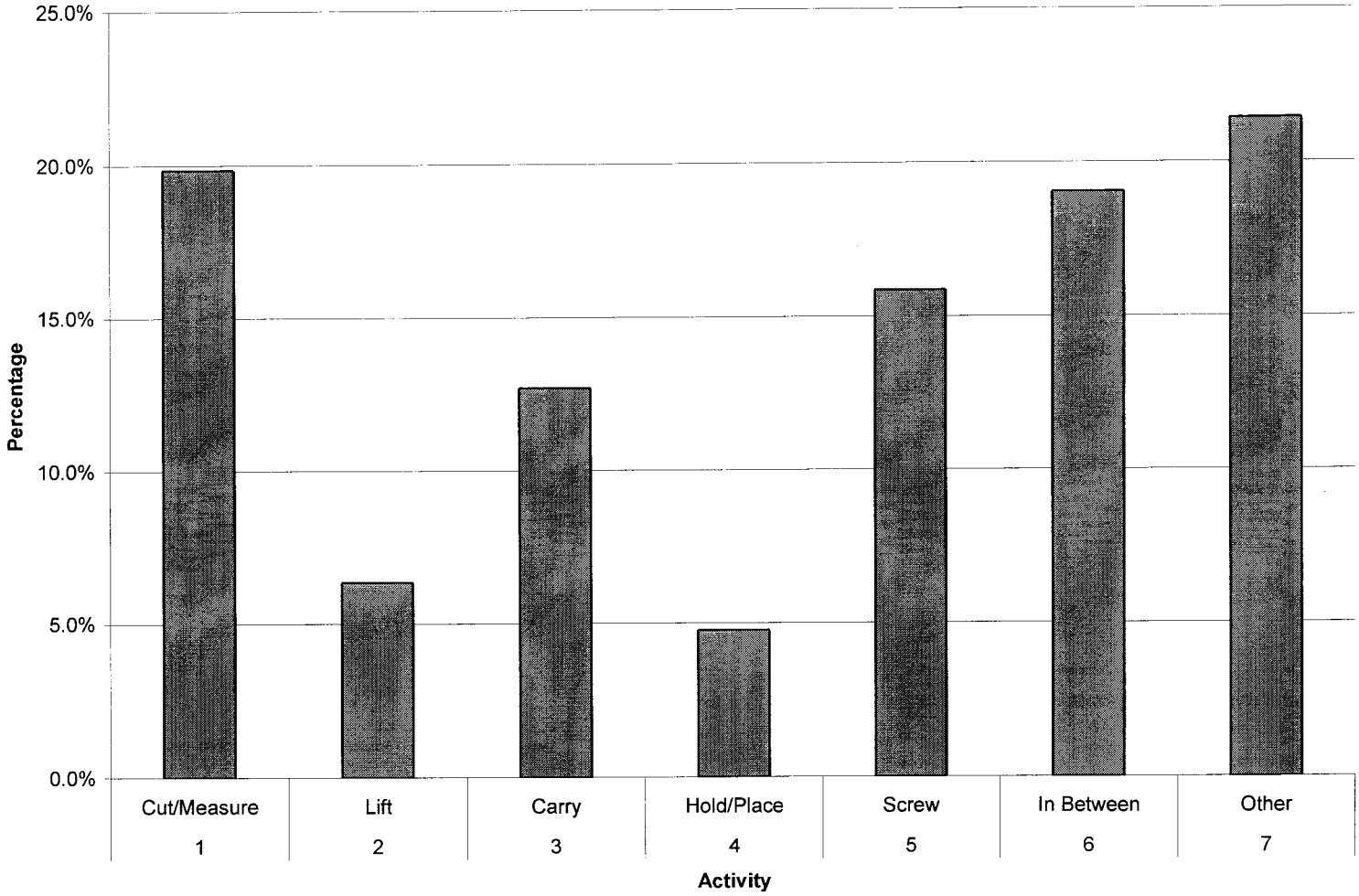


Figure 2 Frequencies of PATH activities

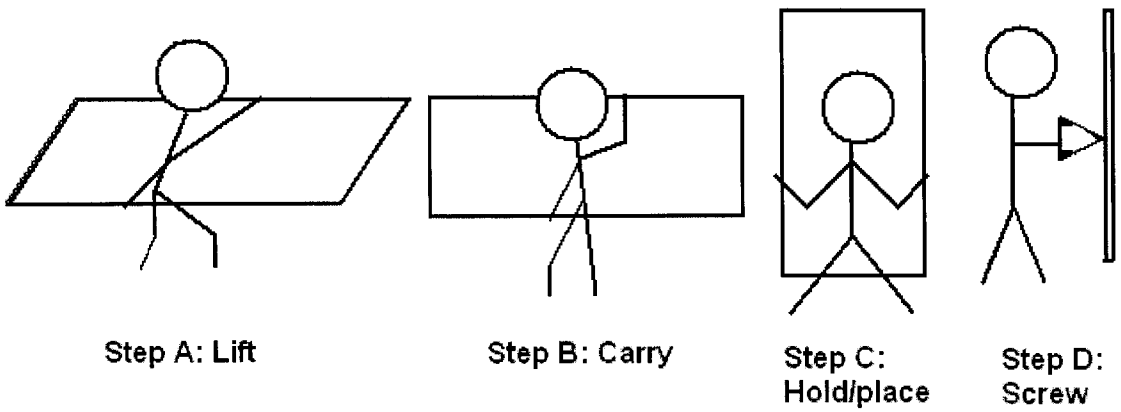


Figure 3 Schematic diagram for installation of a whole drywall sheet

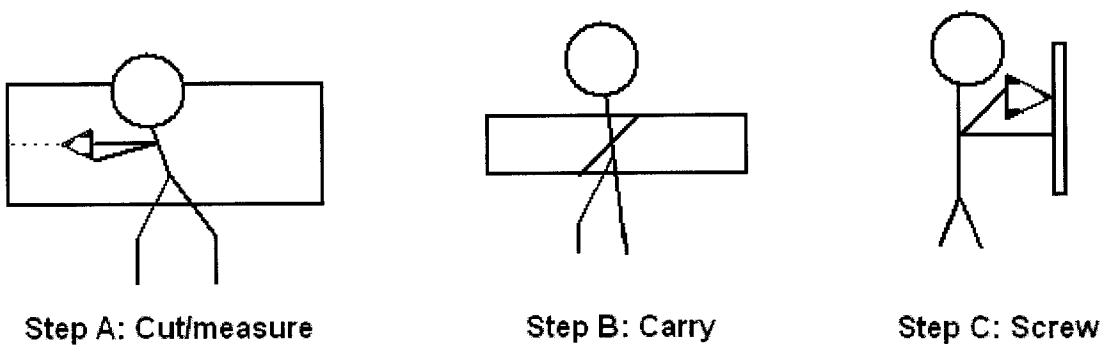
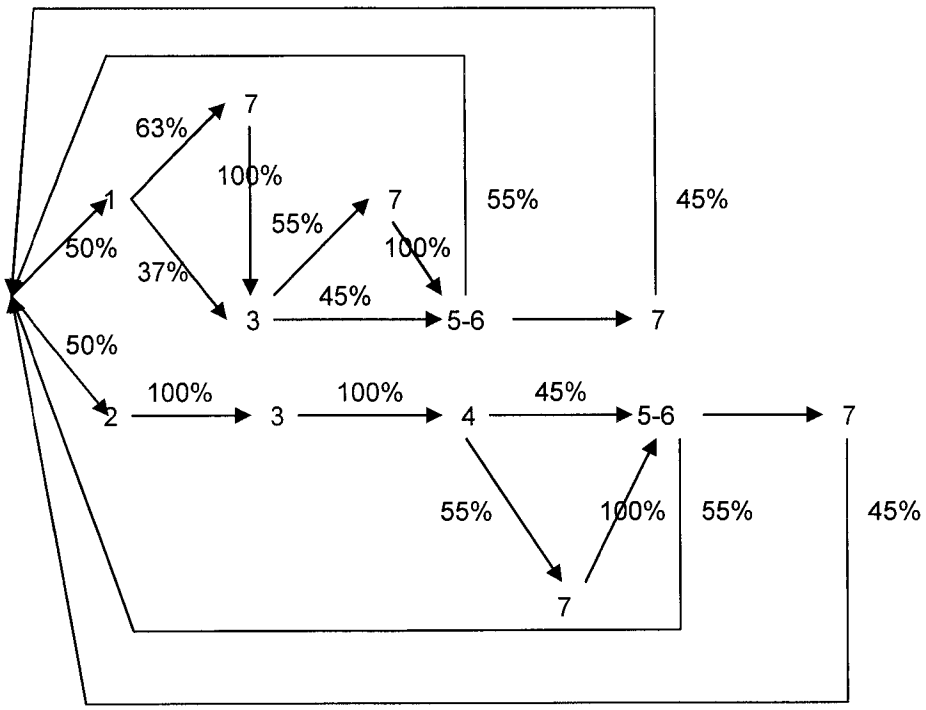
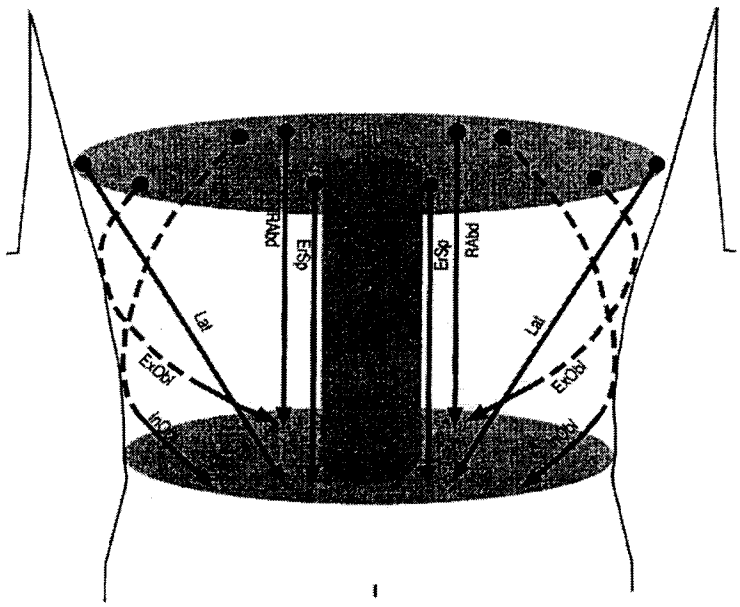


Figure 4 Schematic diagram for installation of a partial drywall sheet



(Activities: 1 – cut/measure, 2 – lift, 3 – carry, 4 – hold/place, 5 – screw, 6 – in between, 7 – other.)

Figure 5 Structure of 8-hour-workday activity series



(Note: ErSp – Erector Spinae, Lat – Latissimus Dorsi, ExObl – External Oblique, InObl – Internal Oblique, Rabd – Rectus Abdominus)

Figure 6 Trunk muscle model developed by Marras and Granata (1995)

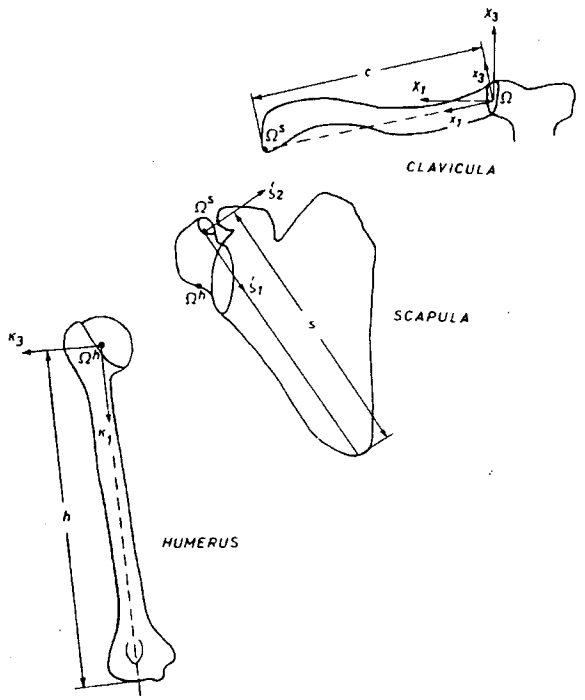
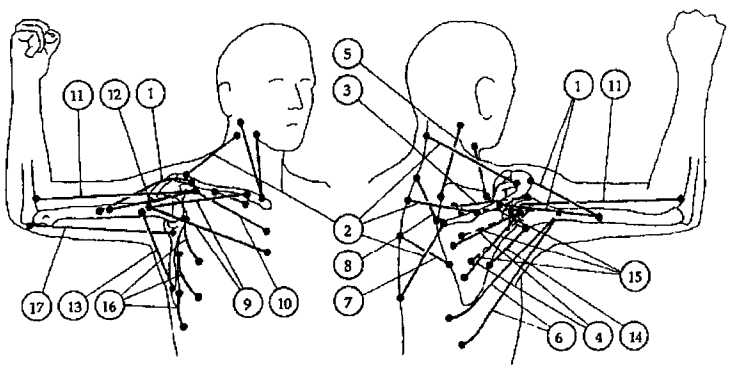


Figure 7 Shoulder model coordinate systems described by Högfors et al. (1987)



(1) deltoideus (three parts)
(2) trapezius (four parts)
(3) supraspinatus
(4) infraspinatus (two parts)
(5) levator scapulae
(6) latissimus dorsi (two parts)
(7) rhomboideus major
(8) rhomboideus minor
(9) pectoralis major (two parts)
(10) pectoralis minor
(11) biceps brachii (two parts)
(12) coracobrachialis
(13) teres major
(14) teres minor
(15) subscapularis (three parts)
(16) serratus anterior (three parts)
(17) triceps (part of)

Figure 8 Shoulder muscle model developed by Högfors et al. (1987) and tested by Karlsson and Peterson (1992)

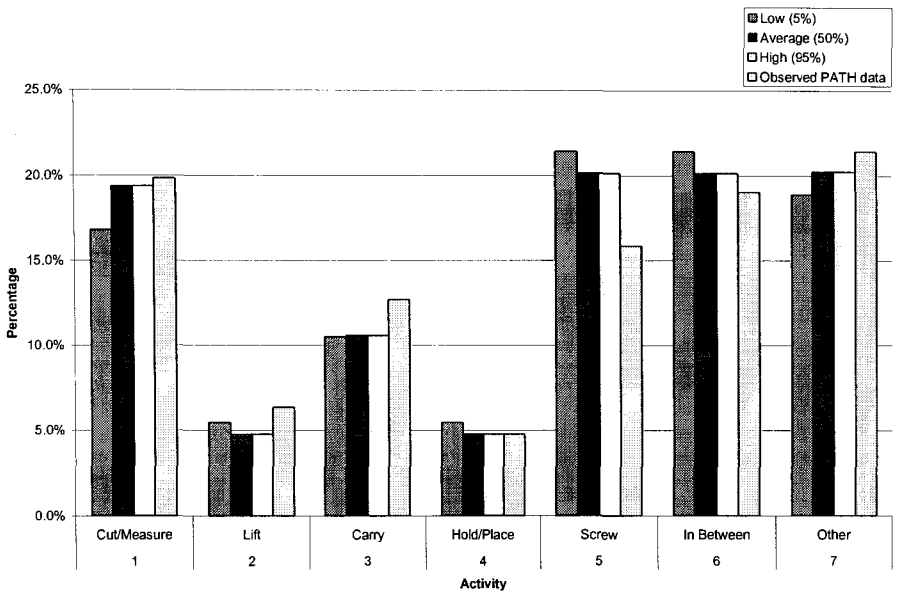


Figure 9 Frequencies of 8-hour-worday activities for 5, 50, and 95 percentiles of the hypothesized population and observed PATH data

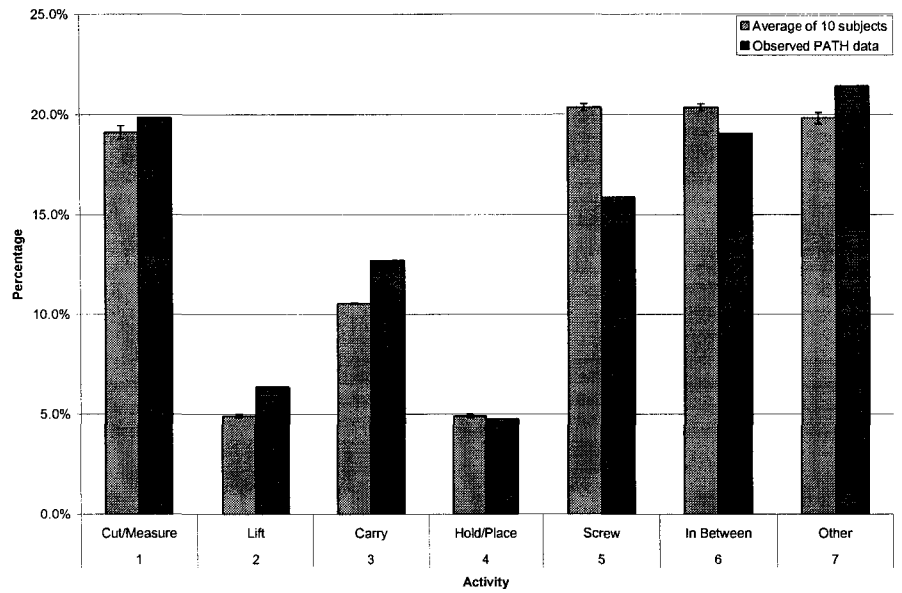


Figure 10 Average of 8-hour-workday activity frequencies for 10 simulated subjects and observed PATH data

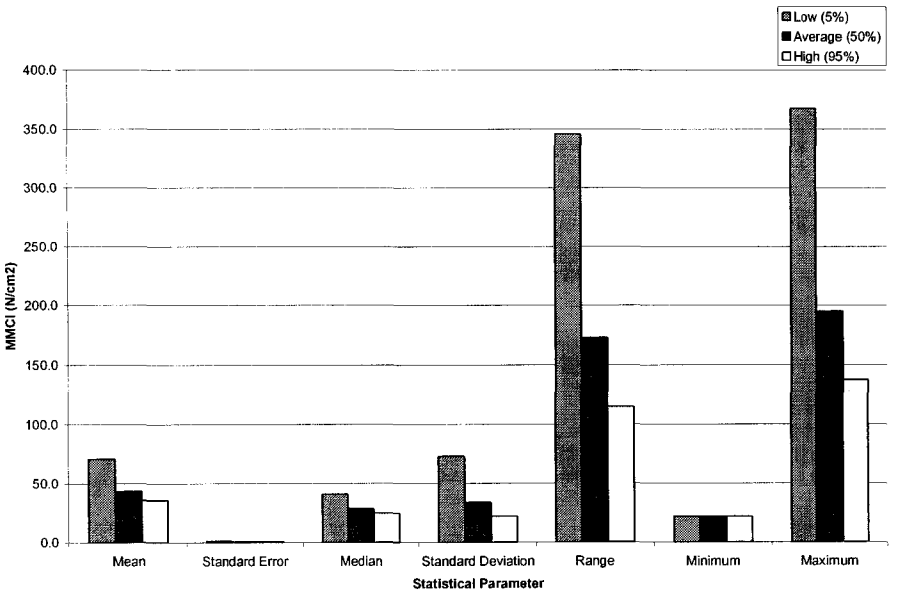


Figure 11 MMCI (Maximum Muscle Contraction Intensity) for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

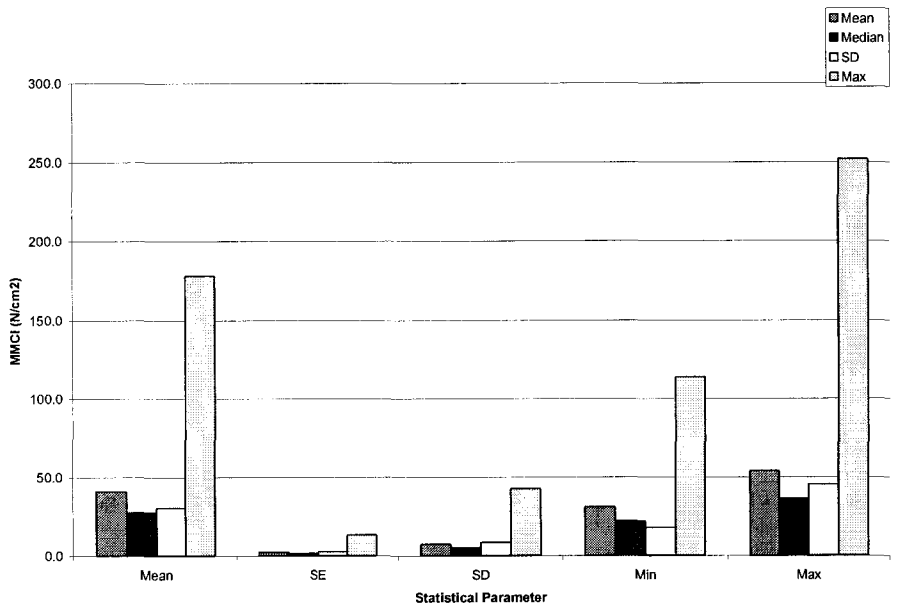


Figure 12 MMCI (Maximum Muscle Contraction Intensity) for 10 simulated subjects during an 8-hour workday

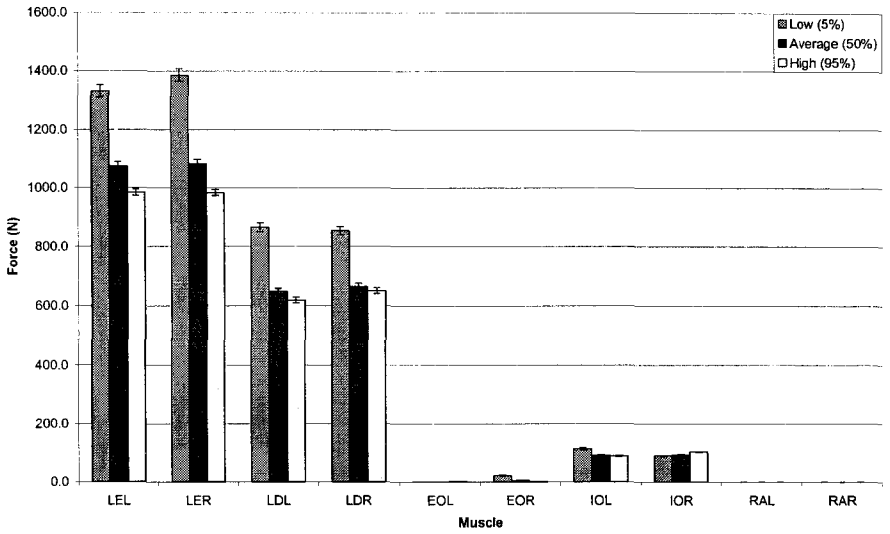


Figure 13 Trunk muscle force means for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

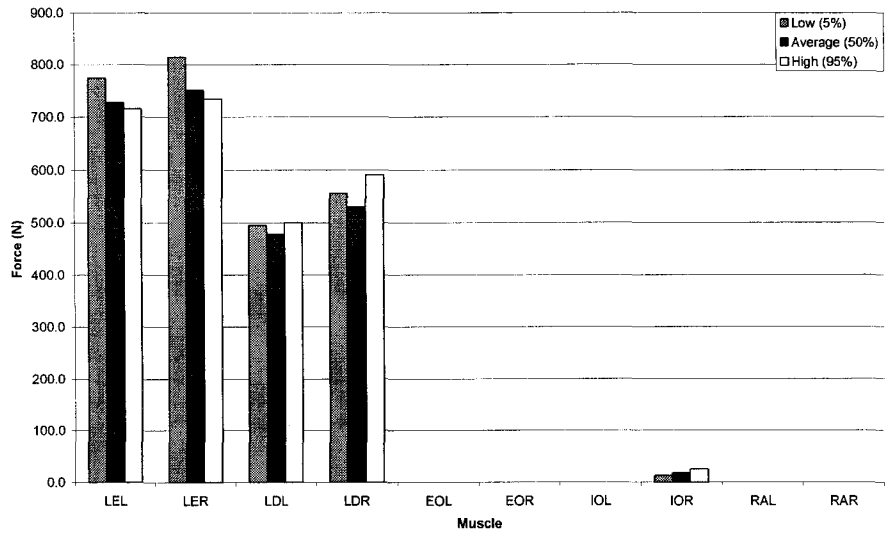


Figure 14 Trunk muscle force medians for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

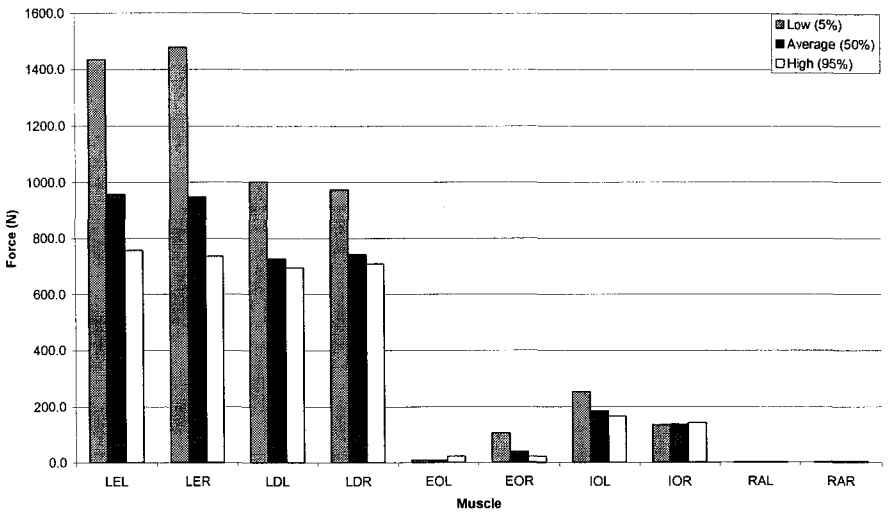


Figure 15 Trunk muscle force standard deviations for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

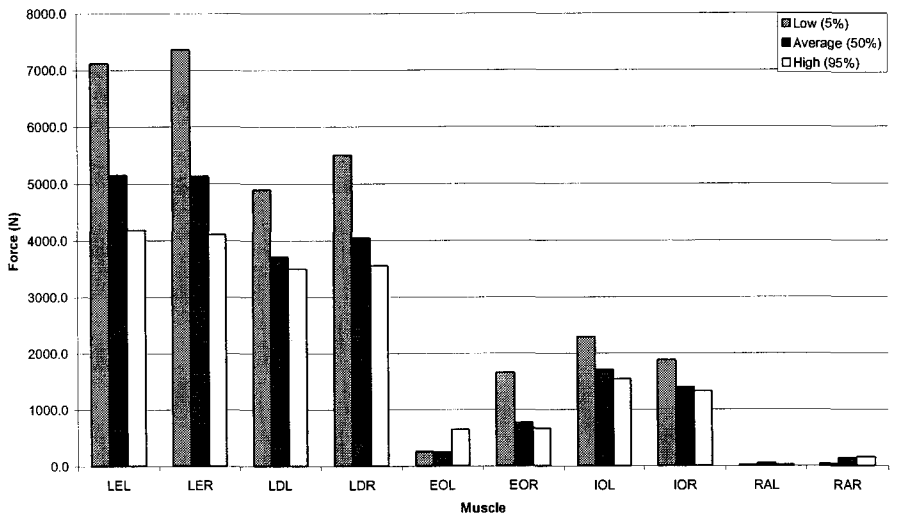


Figure 16 Trunk muscle force maximum values for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

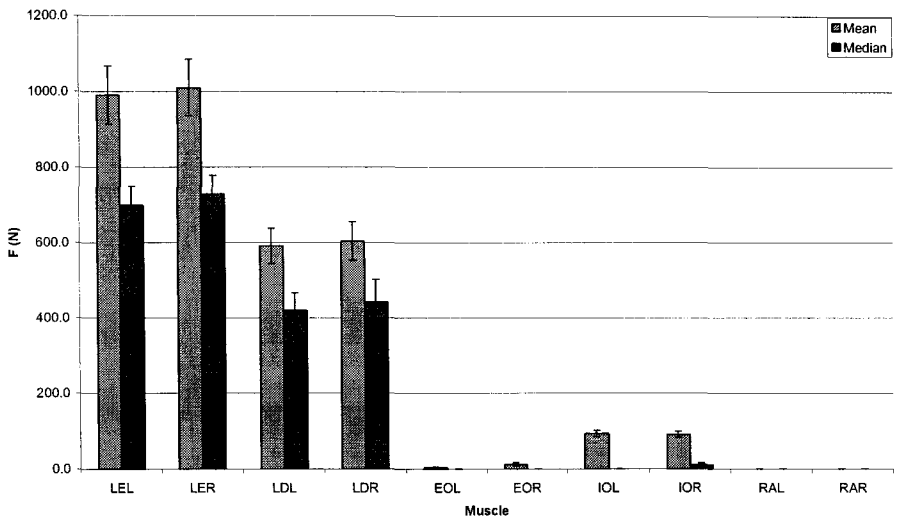


Figure 17 Averages of trunk muscle force means and medians for 10 simulated subjects during an 8-hour workday

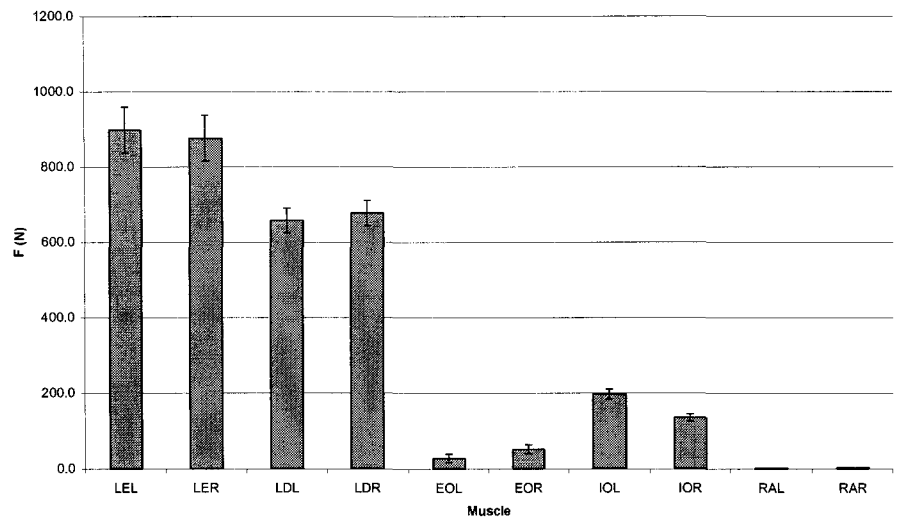


Figure 18 Averages of trunk muscle force standard deviations for 10 simulated subjects during an 8-hour workday

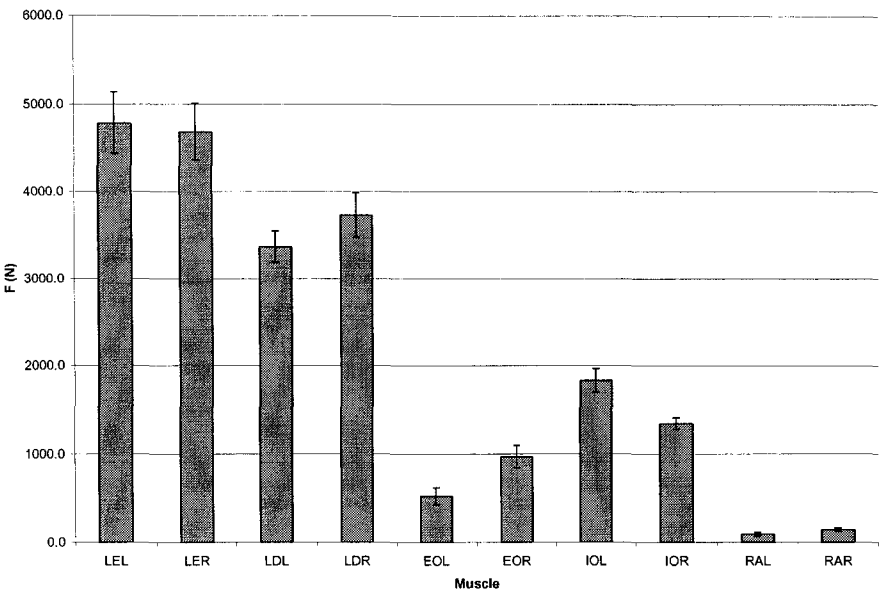


Figure 19 Averages of trunk muscle force maximum values for 10 simulated subjects during an 8-hour workday

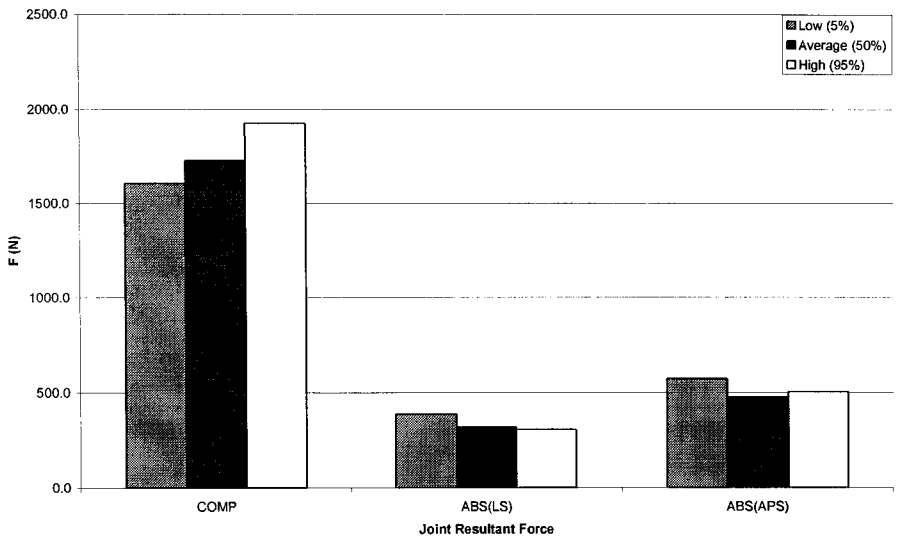


Figure 20 L4/L5 Joint resultant force means for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

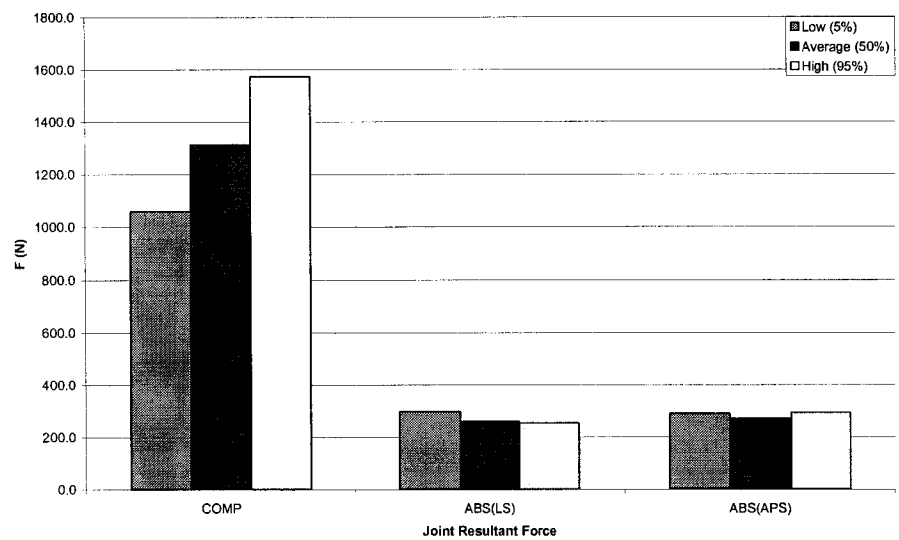


Figure 21 L4/L5 Joint resultant force medians for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

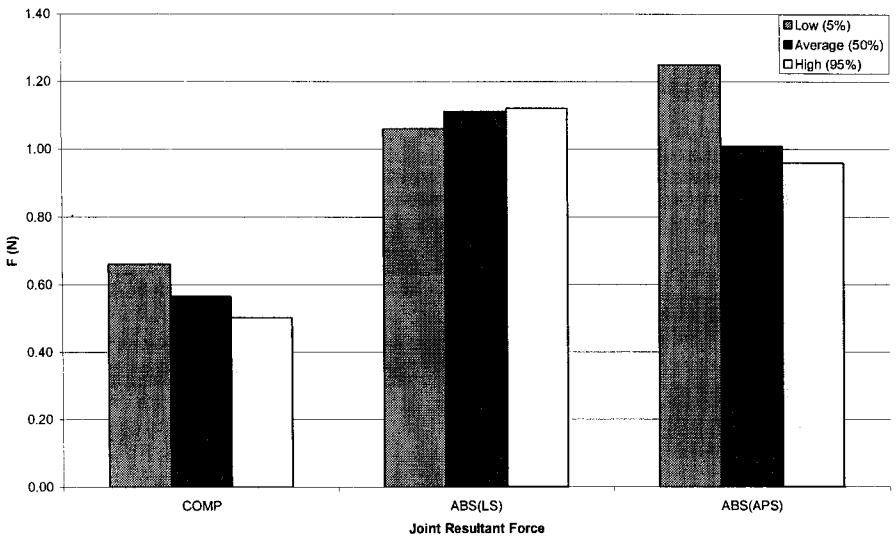


Figure 22 L4/L5 Joint resultant force geometric standard deviations for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

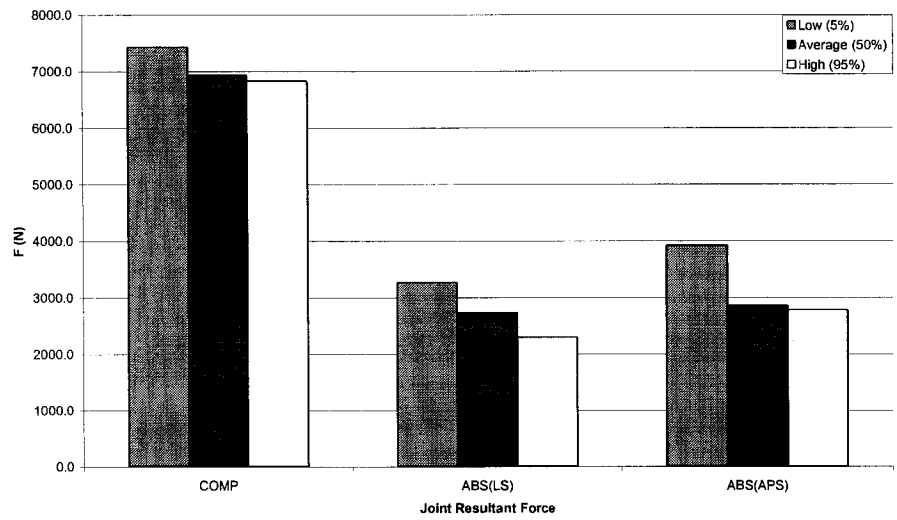


Figure 23 L4/L5 Joint resultant force maximum values for 5, 50, and 95 percentiles of the hypothesized population during an 8-hour workday

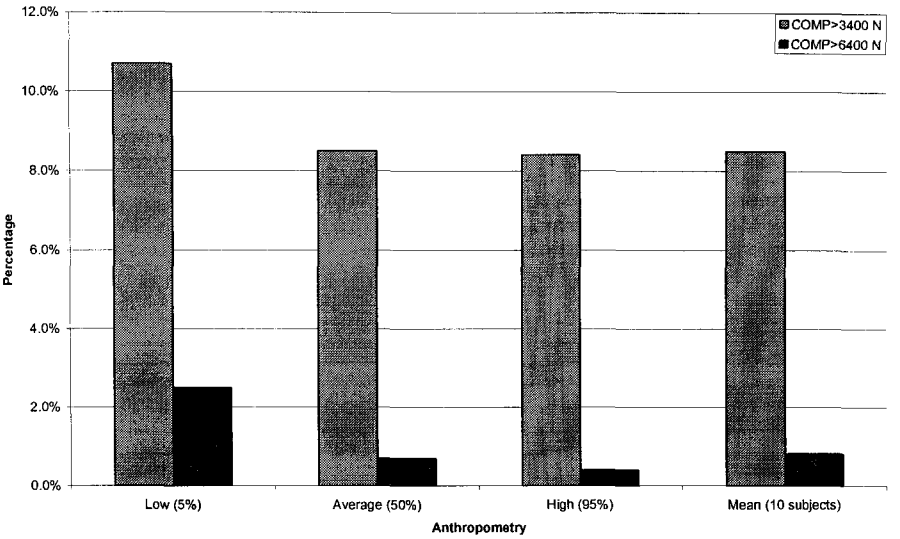


Figure 24 Frequencies of disc compression forces above load tolerances for different anthropometries during an 8-hour workday

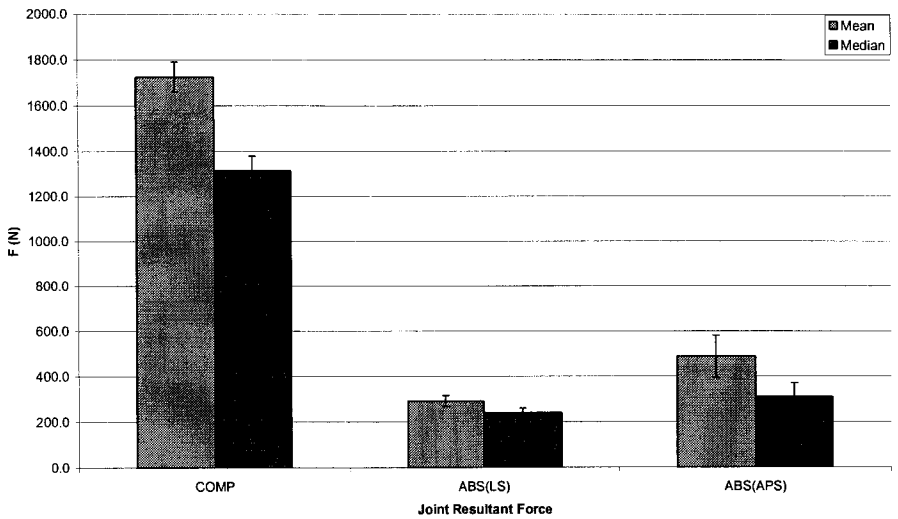


Figure 25 Averages of L4/L5 joint resultant force means and medians for 10 simulated subjects during an 8-hour workday

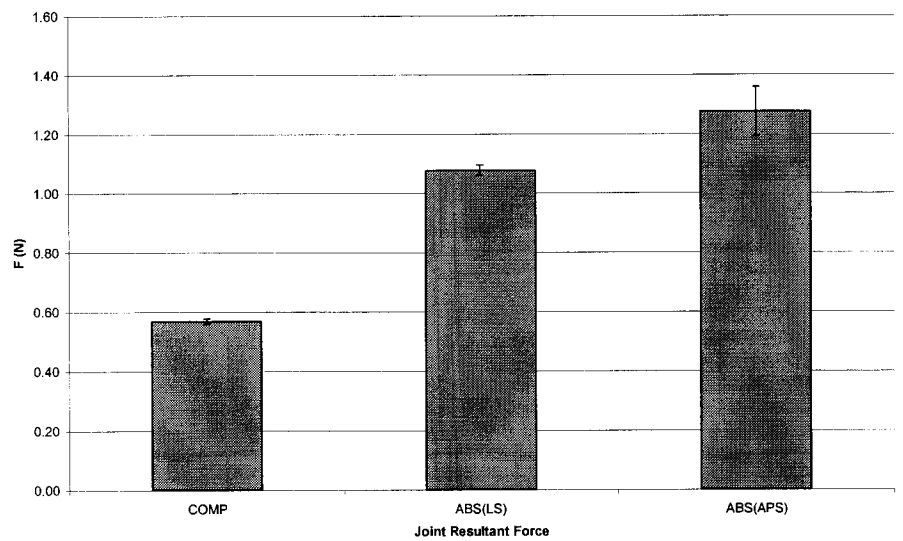


Figure 26 Averages of L4/L5 joint resultant force geometric standard deviations for 10 simulated subjects during an 8-hour workday

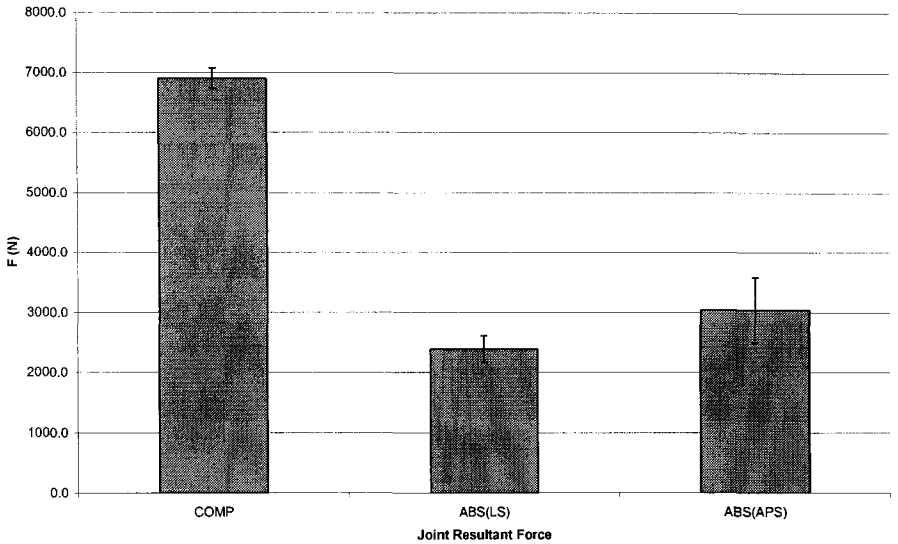


Figure 27 Averages of L4/L5 joint resultant force maximum values for 10 simulated subjects during an 8-hour workday

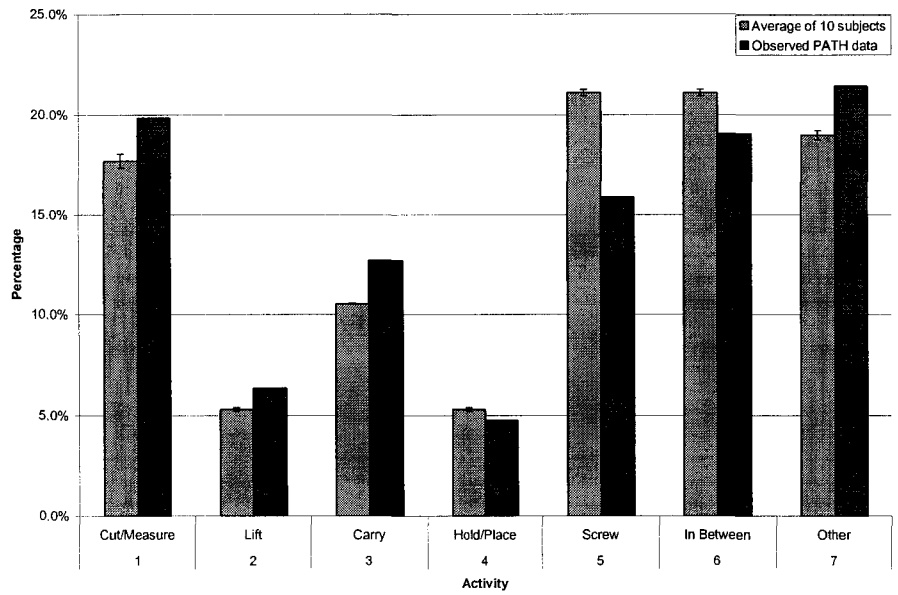


Figure 28 Average of 8-hour-workday activity frequencies for 10 simulated subjects and observed PATH data

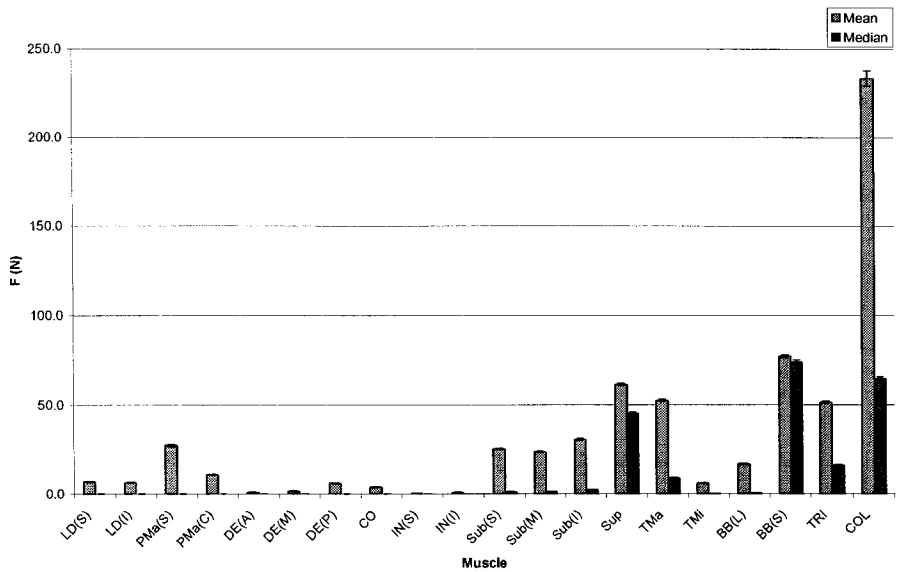


Figure 29 Averages of shoulder muscle force means and medians for 10 simulated subjects during an 8-hour workday

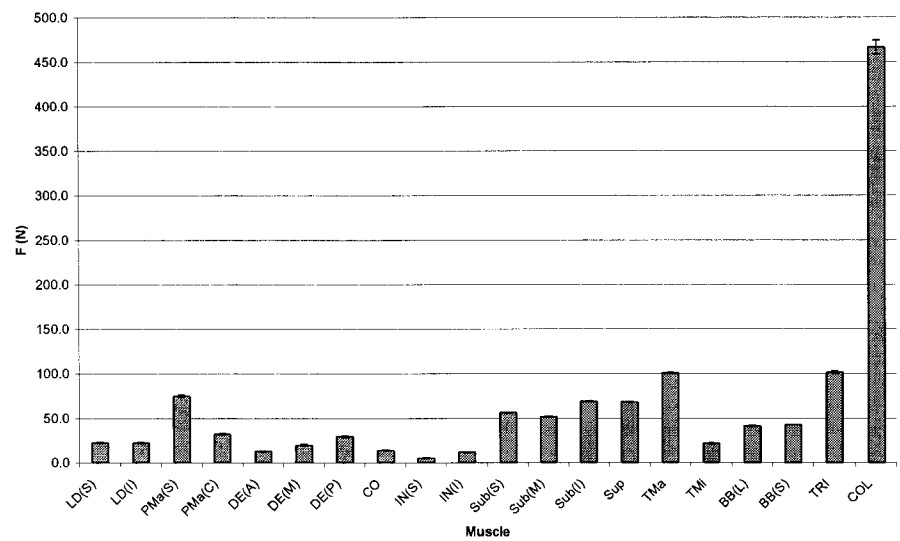


Figure 30 Averages of shoulder muscle force standard deviations for 10 simulated subjects during an 8-hour workday

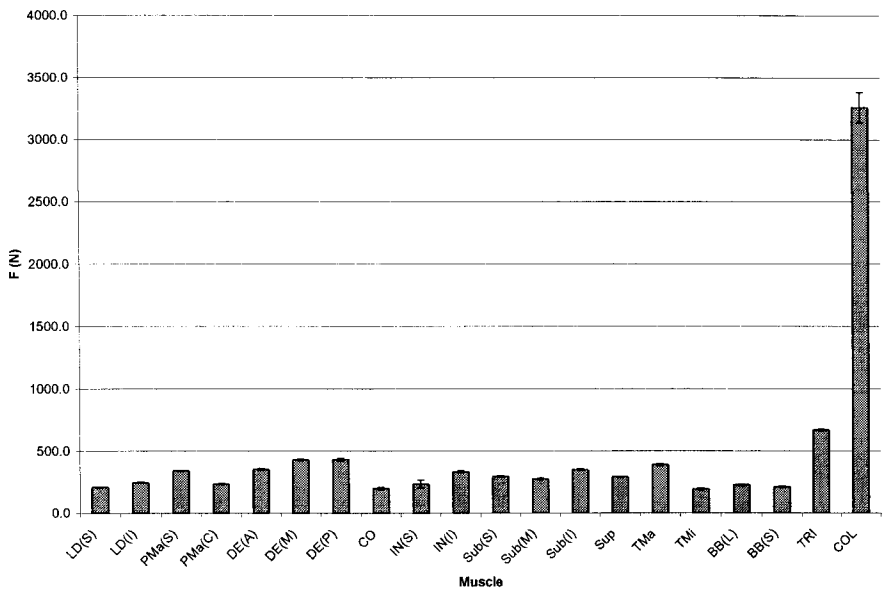


Figure 31 Averages of shoulder muscle force maximum values for 10 simulated subjects during an 8-hour workday

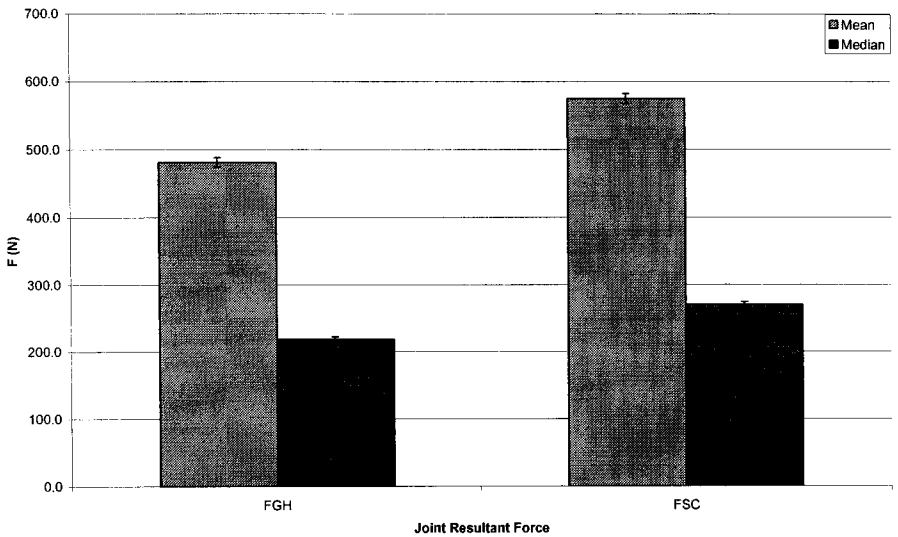


Figure 32 Averages of GH (glenohumeral) and SC (sternoclavicular) joint resultant force means and medians for 10 simulated subjects during an 8-hour workday

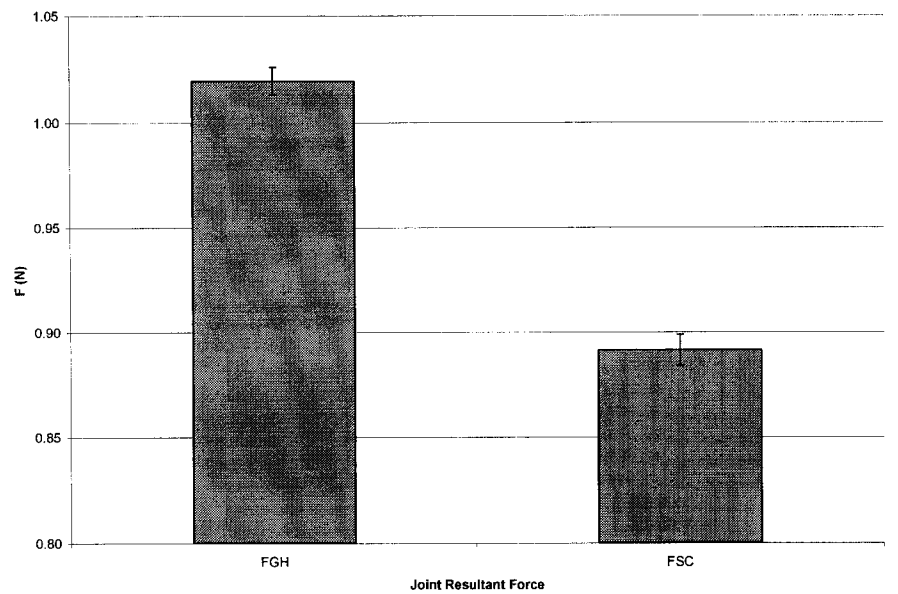


Figure 33 Averages of GH (glenohumeral) and SC (sternoclavicular) joint resultant force geometric standard deviations for 10 simulated subjects during an 8-hour workday

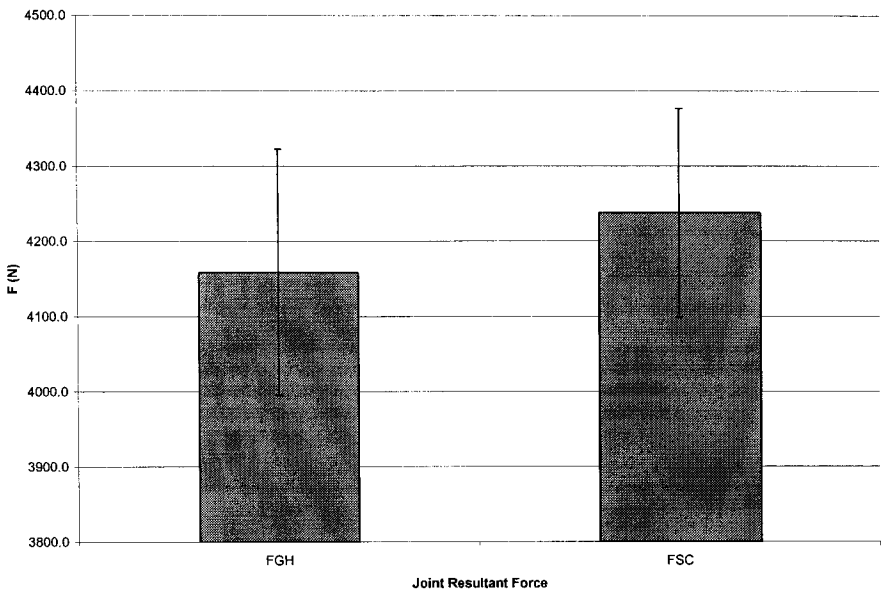


Figure 34 Averages of GH (glenohumeral) and SC (sternoclavicular) joint resultant force maximum values for 10 simulated subjects during an 8-hour workday

III. Examination of muscle fatigue at the low back and shoulder during a simulated workday of drywall installation

A. Introduction

Considering that most construction work requires intense physical effort, it is imagined that muscle fatigue may be a common phenomenon for construction workers. Although the link between muscle fatigue and musculoskeletal injuries and disorders remains unclear (Jørgensen, 1997), it has been hypothesized by several researchers that muscle fatigue may lead to metabolic changes in a muscle that potentially reduces its tolerance to stress (Sjøgaard, 1990; Larsson et al., 1988). Others have suggested that if muscles are required to respond to an unexpected load or a demand that is greater than the compromised muscles' capability, the brunt of the load may be shifted to the passive tissues, and possible injury may result (Parnianpour et al., 1988; Roy, 1993). Muscle fatigue has also been hypothesized to affect neuronal activation patterns and subsequent muscle coordination (Bigland-Richie et al., 1983). All these hypotheses suggest that muscle fatigue is either a contributing factor for or an intermediate process in tissue damage from mechanical overload.

In experimental studies of trunk muscle fatigue, a common pattern is that when the primary agonists fatigue, alternate agonistic and antagonistic muscle groups are recruited increasingly. Sparto et al. (1997) found significant increases in internal oblique and latissimus dorsi muscle activity as the erector spinae became fatigued during isometric trunk extension against varying degrees of resistance. O'Brien and Potvin (1997) observed increased activation of the antagonistic trunk rotators during fatiguing isometric trunk rotation. When it was assumed that the erector spinae were the only

muscle group to fatigue, Sparto and Parnianpour (1998) found increases in the latissimus dorsi and external oblique muscle forces with fatigue, whereas the rectus abdominus and internal oblique showed variable changes with fatigue. Furthermore, significant increases in anteroposterior shear and compression force were also detected using the EMG-assisted models of spinal loading.

Similarly, normal shoulder function is dependent on a combination of intact static and dynamic stabilizers. The function of the dynamic stabilizers seems to be dependent on fatigue. The rotator cuff muscles, consisting of supraspinatus, infraspinatus, subscapularis, and teres minor, maintain the humeral head relative to the glenoid socket and supply force to move the arm in an arc of abduction and rotation. It is likely that the centering effect of the rotator cuff was decreased with fatigue (Chen et al., 1999). Tsai et al. (2003) considered the infraspinatus and teres minor muscles as the primary external rotators of the glenohumeral joint and suggested a decrease of more than 25% of external rotator torque as the criterion for muscle fatigue. Côté et al. (2005) conducted a comparative experimental study of repetitive hammering between healthy, fatigued, and shoulder-injured individuals, and found that when fatigued, the healthy subject displayed decreased range of joint motion, peak velocity and peak acceleration of elbow motion as well as reduced grip strength.

A variety of experimentally derived techniques have been developed to quantify muscle fatigue and provide guideline for acceptable work-rest schedules, focusing on different aspects of psychophysical and physiological discomfort and muscular endurance (Von Rohmert, 1973; Dul et al., 1991; Fisher et al., 1993; Wood et al., 1997; Lin and Radwin, 1997). With the general goal of biomechanical analysis of drywall installation,

an approach using muscular endurance was considered based on the challenge-recovery model of muscle fatigue. As suggested by Freund and Takala (2001), muscle fatigue is accounted for by introducing force capacity consumption and recovery mechanisms so that the force production capacity varies between zero and the upper limits of the muscle forces. The muscle is a kind of reservoir whose force production capacity reduces with the time that the muscle is contracted. At the same time, the reservoir is filled with more capacity by the surrounding systems.

The objective of this study was to examine the potential muscle fatigue for drywall installers during a simulated 8-hour workday, using the one-compartment challenge-recovery model. The required muscle contraction forces and joint resultant forces at the low back and shoulder were first estimated by integrating the methodologies of PATH, Monte-Carlo simulation, and biomechanical modeling. Specific muscle groups that are considered to fatigue were then identified and their upper limits of contraction forces, recovery rate and decay rate were either assumed or calculated. In the end, muscle fatigue rates and consecutive fatigue periods were defined and computed to illustrate the cumulative effects of muscle fatigue on the physical requirements of drywall installation.

B. Methods

1. Determination of required muscle contraction forces

The study explores an integrating approach through which the required muscle contraction forces and joint resultant forces at the low back and shoulder during drywall installation were determined (Figure 35). For detailed description of the methods, please refer to Chapter II of this dissertation (Yuan, 2006). PATH (Posture, Activity, Tools, and Handling), the observational work-sampling based approach developed by Buchholz et al. (1996), was used to quantify the percent of time that the drywall installers were conducting different activities with different body part postures. An 8-hour-workday activity series was simulated using the Monte-Carlo simulation based on activity percentages in the PATH data (Step A in Figure 35). The biomechanical models of the low back developed by Marras and Granata (1995) and the shoulder model presented by Högfors et al. (1987) and tested by Karlsson and Peterson (1992) were utilized in the study to determine the musculature-associated parameters. The model input variables, including the anthropometric data, the joint angles, the external load force and position vectors, and the internal muscle parameters, were generated by Monte-Carlo simulation for the analyses of the low back and shoulder (Steps B1 and B2 respectively in Figure 35). Utilizing different optimization programs in MATLAB (The MathWorks, Natick, MA, U.S.A.), the three-dimensional static equilibrium equations were solved and the biomechanical model output variables of muscle contraction forces and joint resultant forces at the low back and shoulder for both the selected 5, 50, and 95 percentiles and 10 randomly generated subjects from the hypothesized population (Marras et al., 2001) were obtained (Steps C1 and C2 respectively in Figure 35).

2. One compartment model

The original equation designated to examine muscle fatigue that was proposed by Freund and Takala (2001) was:

$$\frac{d}{dt}(S^0) = \mu * (S^m - S^0) - \nu * S = (\mu * S^m - \nu * S) - \mu * S^0 \quad (3.1)$$

Where,
S^m = The upper limit of muscle contraction force
S^0 = Muscle force production capacity, which varies between zero and S^m
S = Muscle contraction force
μ = Muscle recovery rate
ν = Muscle decay rate

An approximate difference equation for this model is:

$$S^0_{t+1} = S^0_t + \Delta t * ((\mu * S^m - \nu * S_t) - \mu * S^0_t) \quad (3.2)$$

Where,
S_t = Muscle contraction force at t, obtained from MATLAB
S^0_t = Muscle force production capacity at t
S^0_{t+1} = Muscle force production capacity at t+1
Δt = Time interval (0.1 min)

If $S^0_t < S_t$, the muscle cannot generate the required contraction forces and needs supplement from other muscle groups. A muscle is then defined as fatigued at t.

A common work-rest schedule during an 8-hour workday in the construction industry, as frequently observed in the field, is to start around 6 o'clock in the morning, and then get a 15-min break around 9 o'clock, a 30-min lunch break during noontime, and a 15-min break in the afternoon. The maximum continuous work period is 3 hours. During the breaks, the muscles rest and their contraction forces were assumed to be zero.

Muscle force production capacity could then be regenerated and muscle fatigue would be reduced during the break.

3. Muscles considered to fatigue

Many researchers assume the lumbar erector spinae to be the only muscle group to fatigue in the trunk. Since there was a pair of erector spinae involved, the higher value of muscle contraction forces of LEL and LER was chosen as the model input.

In terms of the shoulder, significant emphasis has been put on the rotator cuff muscles during examination of muscle fatigue. On the other hand, rotator cuff damage, such as impingement syndrome and tears, is one of the most common shoulder injuries in many occupational settings. Rotator cuff muscles were thus examined in this study. Specifically, muscle contraction forces of different parts of infraspinatus, subscapularis, supraspinatus, and teres minor were summed to represent the input for the one compartment model.

4. Determination of S^m, μ, and ν

Farfan (1973) estimated a normal range for the strength capability of the erector spinae in a healthy population to be 2200-5500 N. Considering that construction workers tend to be stronger generally due to the work involved, the upper limit of erector spinae contraction force was assumed to be 6500 N.

The upper limit of rotator cuff contraction force was calculated using the following equation:

$$F_{\max} = k * (PCSA_{Sup} + PCSA_{IN} + PCSA_{Sub} + PCSA_{TMi}) \tag{3.3}$$

Here k is a constant which depends on the maximum tension in muscles. Its value has been suggested to be in the range 40-100 N/cm^2 (Crowninshield and Brand, 1981). According to Wood et al. (1989), the average maximum muscle stress is about 100 psi (pounds per square inch), which is equivalent to 70 N/cm^2. The PCSAs (Physiological Cross-Sectional Area) of rotator cuff muscle components were obtained from Karlsson and Peterson (1992). As a simple estimate, the upper limit of rotator cuff muscle force was assumed to be 1500 N.

In order to determine the muscle recovery rate and decay rate, the relationship between endurance time and muscle contraction force level needs to be determined. The Rohmert curve (Von Rohmert, 1960) established such a hyperbolic relationship (Figure 36). The endurance time was defined as a time taken for a muscle to lose its volitional, sustained force-producing capability, and was considered primarily a function of the contraction force, the exercise-rest cycle, and the composition of fibers within the muscle (Chaffin et al., 1999).

Specifically, shoulder muscle recovery rate and decay rate were determined by Freund and Takala (2001) using the equation:

$$t = -\frac{\ln[1 + \frac{\mu}{\nu}(1 - \frac{S^m}{S})]}{\mu} \tag{3.4}$$

Data on the trunk muscle endurance time and muscle exertion level (percent of maximum voluntary contraction) were obtained from Van Dieën et al. (1998), where the endurance times were 10.1, 2.4, 0.9 min, for the 25%, 50% and 75% MVC condition, respectively. A curve-fitting program was then executed in MATLAB to solve the equation (3.4) for μ and ν of the erector spinae muscle (Appendix D).

Table 25 summarizes the muscle fatigue variables included in the model.

5. Model output variables

5.1. Prevalence of muscle fatigue

The prevalence of muscle fatigue during the simulated 8-hour workday was computed using:

$$P = \frac{Time\ Units\ (Muscle\ Fatigue)}{Total\ Time\ Units} * 100\% \tag{3.5}$$

Where:
P = Prevalence of muscle fatigue
Time Units (*Muscle Fatigue*) = Number of time units when muscles are considered fatigued according to the model definition
Total Time Units = Number of the total time units during the simulated workday

5.2. Consecutive fatigue periods

Using equation (3.4), the endurance times of 50% MVC of the lumbar erector spinae and the rotator cuff muscles were found to be 2.4 mins and 1.0 mins respectively. The consecutive fatigue period was defined as a period lasting at least 2.4 mins and 1.0 mins for the trunk and shoulder, respectively, when muscles were continuously fatiguing. The average durations (AD) of consecutive fatigue periods during the simulated workday for different subjects were also calculated by:

$$AD = \frac{Sum\ of\ durations\ of\ consecutive\ fatigue\ periods}{Number\ of\ consecutive\ fatigue\ periods} \tag{3.6}$$

6. Data analysis

The prevalences of trunk and shoulder muscle fatigue during the simulated 8-hour workday were calculated for different subjects selected in the study. Frequencies of simultaneous muscle fatigue and joint resultant forces above the load tolerances were also computed. The total count of consecutive fatigue periods for the trunk and shoulder muscles during the simulated workday were obtained, as well as their average durations.

C. Results

1. Prevalences of muscle fatigue

The prevalences of trunk muscle fatigue were 57.3%, 40.4%, and 27.9%, for the selected 5, 50, and 95 percentiles of the hypothesized population, respectively. The average of trunk muscle fatigue frequencies for the 10 randomly generated subjects was 33.5% with a high standard deviation of 28.8%. The highest frequency was 74.5%, which indicated that the 8-hour workday required strenuous effort for such a subject based on the fatigue definition in the study (Table 26).

When associated with the scenario of disc compression forces above load tolerances (3400 N and 6400 N), frequencies of simultaneous muscle fatigue and disc compression forces above the load tolerances were very close to the frequencies of disc compression forces above the load tolerances only (Figure 37). Such results have potentially produced evidence to link trunk muscle fatigue and overload of the lumbar spine.

The rotator cuff muscle fatigue prevalence was 5.1% on average for 10 randomly simulated subjects. The standard deviation was 0.8%, which suggests low inter-subject variability (Table 27).

Using the body weight (BW) as a proposed reference for shoulder load tolerance, the average frequencies of simultaneous rotator cuff muscle fatigue and the GH and SC joint resultant forces above the load tolerance were 5.1% (SD 0.7%) and 5.2% (SD 0.7%), respectively. Meanwhile, the frequencies of joint resultant forces above the load tolerance alone were 16.8% (SD 1.3%) and 18.4% (SD 1.3%), respectively (Figure 38). Such differences might indicate that more muscle groups besides the rotator cuff should be

considered in terms of linking fatigue to overload of the shoulder joints, if the suggested shoulder load tolerance is reasonable.

2. Consecutive fatigue periods

The total number of consecutive fatigue period during the simulated workday was 47, 29, and 9 for the selected 5, 50, and 95 percentiles of hypothesized population, respectively. The average durations of consecutive fatigue periods for those different anthropometries were 3.6, 3.4, and 3.3 min, respectively. There were 15 (SD 17) consecutive fatigue periods on average for the 10 randomly generated subjects and the average duration was 3.7 (SD 1.4) min. The maximum count of consecutive fatigue periods was 39 and the longest average duration was 5.8 min.

No consecutive fatigue period of the rotator cuff muscles was found, which potentially indicated that shoulder muscle fatigue tended to be intermittent during the simulated 8-hour workday according to the fatigue definition by the one-compartment model.

D. Discussion

Using the one-compartment challenge-recovery model, this study examined potential muscle fatigue at the lumbar spine (L4/L5) and shoulder during a simulated 8-hour drywall installation work. The prevalences of muscle fatigue and the number and average duration of consecutive fatigue periods were computed to describe both the cumulative and concentrated effects of muscle fatigue on the physical requirements for the drywall installers. It was found that the trunk muscles were much more vulnerable to fatigue, compared to the shoulder muscles. This was in agreement with the results from equation (3.4) that the trunk muscle recovery rate was almost half of the shoulder muscle recovery rate, given that the decay rate had less influence on the muscle fatigue which was determined by the equation.

The relationship between muscle fatigue and overload at the trunk and shoulder was also investigated by computing the frequencies of simultaneous muscle fatigue and overload. The results suggested a good linkage between trunk muscle fatigue and disc overload; however, no strong evidence between rotator cuff muscle fatigue and overload at the shoulder joints could be provided by this study. On the one hand, the rotator cuff may not be the only shoulder muscle group to fatigue; on the other hand, the shoulder load tolerance value used here has not been widely accepted.

Subject body weight (BW) was selected as the shoulder load tolerance value based on previous studies (Poppen and Walker, 1978; Karlsson and Peterson, 1992; Anglin et al., 2000). It should be mentioned that different from the above studies which usually evaluated the joint resultant forces as how many times of BW, this study focused

on calculating the frequencies of joint resultant forces exceeding BW and also used those variables to examine the linkage between muscle fatigue and joint overload.

Although the relationship between muscle fatigue and musculoskeletal injuries remains undefined, the idea that reasonable work-rest schedules could minimize muscle fatigue is popular. Muscle fatigue frequency could then be considered a tool for evaluating the acceptableness of a work-rest schedule.

In this study, the endurance times for 50% MVC of the trunk and shoulder muscles were selected as criteria to determine the consecutive fatigue periods. This idea comes from the definition of "half life", which is usually examined in a one-compartment model. Similar to how half life represents the half time of clearance or repair, the duration of a consecutive fatigue period is when the subject is maintaining half MVC. Of course, there is lack of information to define how long a continuous fatigue period is too long for a person. Thus, the introduction of consecutive fatigue period was aimed as a relative comparison between pre- and post- ergonomic interventions, as it was hypothesized that while a certain intervention might reduce the muscle fatigue frequency cumulatively, the concentrated consecutive fatigue periods might be increased instead.

Given that the drywall installation task, especially the included activity of lifting, usually requires significant trunk muscle forces, the upper limit of lumbar erector spinae force was assumed to be 6500 N. Such a value is much higher than the normal range from 2200 N to 5500 N (Farfan, 1973), which includes both the left and right side of lumbar erector spinae. However, as Holley et al. (1999) found that the lumbar erector spinae forces actually exceeded such a range during a simulated sailboarding, it is surmised that

the selected upper limit may be reasonable during lifting of the heavy and bulky drywall sheets.

Since the PCSA of the lumbar erector spinae for the average person from the hypothesized population in this study was about 27 cm^2, the maximum stress for the erector spinae muscle would be more than 200 N/cm^2 under the assumption that the highest erector spinae muscle force equaled 6500 N. This is much higher than the assumed value for the rotator cuff muscles (70 N/cm^2). Buchanan (1995) concluded that the maximum muscle stress was not a constant based on the differences in specific tension in elbow flexors and extensors, so it may be reasonable that trunk and shoulder muscles can endure different maximum stresses too. On the other hand, even the maximum lumbar spinae muscle force and stress were assumed to be higher than the normal range, the prevalence of trunk muscle fatigue under the definition in this study was still much higher than that of shoulder muscle fatigue. This emphasizes the concern that the trunk muscles are more vulnerable to fatigue than the shoulder muscles during drywall installation.

It should be recognized that the one-compartment challenge-recovery model usually requires a slow change of the input values along time. In this study, the model input was computed as $\mu * S^m - \nu * S_t$, which was determined by the upper limit of the muscle contraction force (S^m), muscle recovery (μ) and decay rate (ν), and the required muscle contraction forces (S_t) obtained from the biomechanical models. The selection of small time interval (0.1 min) as commonly used in the one-compartment models might have helped reduce the pace of changes in the model input. However, since the required muscle forces for conducting different activities varied dramatically, the input values for

the one-compartment model still changed very quickly. This might have compromised the validity of results. Ideally, had smaller time interval been used (for example, 0.01 min), there might have been much slower changes in the model input and consequently it might have enhanced the validity of muscle fatigue results; however, on the other hand, it would likely have cost much longer time to run the programs in MATLAB in order to get the muscle force results.

In addition, the selections of erector spinae and rotator cuff as the muscles to fatigue at the low back and shoulder were based on previous literature without perfect evidence. In reality, the fatiguing muscles may vary remarkably depending on many different factors such as activities conducted, postures posed, and work environment exposed, etc. Subjects' individual characteristics may also produce influence on both which muscle group fatigue and how. The variability of muscle fatigue thus, warrants further investigation.

E. Conclusions

Trunk muscles were found more vulnerable to fatigue compared to shoulder muscles during the simulated 8-hour drywall installation work using a one-compartment challenge-recovery model. The prevalences of muscle fatigue, the counts and average durations of consecutive fatigue period could be regarded as proxies for the evaluation of reasonable work-rest schedule as one kind of ergonomic intervention strategy.

F. Literature Cited

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Table 25

Muscle fatigue variables in the one-compartment model

Body Part	Muscles	Input	S^m (N)	μ (min^{-1})	v (min^{-1})
Trunk	Lumbar Erector Spinae	Higher value of LEL and LER	6500*	0.09	0.46
Shoulder	Rotator Cuff	Sup+IN+Sub+TMi	1500**	0.20	1.13

Note:

* – Assumed value based on Farfan (1973); ** – Assumed value based on equation (3.3);

LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; Sup – Supraspinatus; IN – Infraspinatus; Sub – Subscapularis; TMi – Teres Minor; S^m – The upper limit of muscle contraction force; μ – Muscle recovery rate; v – Muscle decay rate.

Table 26

Trunk muscle fatigue for 5, 50, and 95 percentiles of the hypothesized population and 10 simulated subjects during an 8-hour workday

Subject		Fatigue Frequency	Consecutive Fatigue Period		COMP > 3400 N	COMP > 3400 N & Muscle Fatigue	COMP > 6400 N	COMP > 6400 N & Muscle Fatigue
			Number	Average Duration (min)				
Low (5%)		57.3%	47	3.6	10.7%	10.7%	2.5%	2.5%
Average (50%)		40.4%	29	3.4	8.5%	8.4%	0.7%	0.7%
High (95%)		27.9%	9	3.3	8.4%	8.2%	0.4%	0.4%
10 Subjects	Mean	33.5%	15	3.7	8.5%	7.9%	0.8%	0.8%
	SD	28.8%	17	1.4	1.2%	1.8%	1.0%	0.9%
	Max	74.5%	39	5.8	10.9%	10.7%	3.0%	2.9%

Note: COMP – Disc compression force.

Table 27

Shoulder muscle fatigue for 10 simulated subjects during an 8-hour workday

Parameter	Fatigue Frequency	Consecutive Fatigue Period	FGH>BW	FGH>BW & Muscle Fatigue	FSC>BW	FSC>BW & Muscle Fatigue
Mean	5.1%	0	16.8%	5.1%	18.4%	5.2%
SD	0.8%	0	1.3%	0.7%	1.3%	0.7%
Max	6.3%	0	19.3%	6.1%	20.9%	6.1%

Note: FGH – Joint resultant force at glenohumeral; FSC – Joint resultant force at sternoclavicular; BW – Body weight.

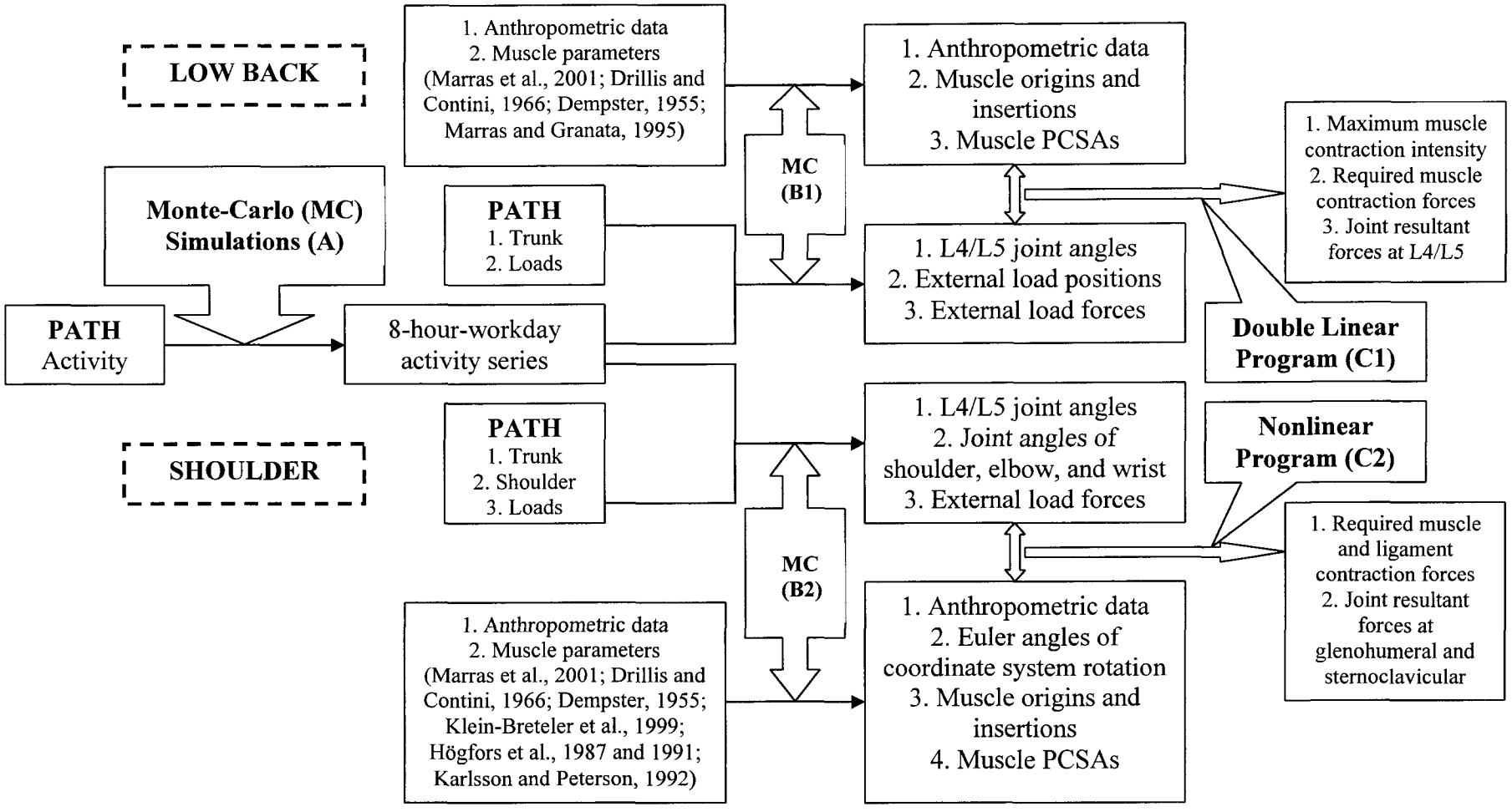

Figure 35 Schematic diagram for determination of muscle contraction forces and joint resultant forces in the study

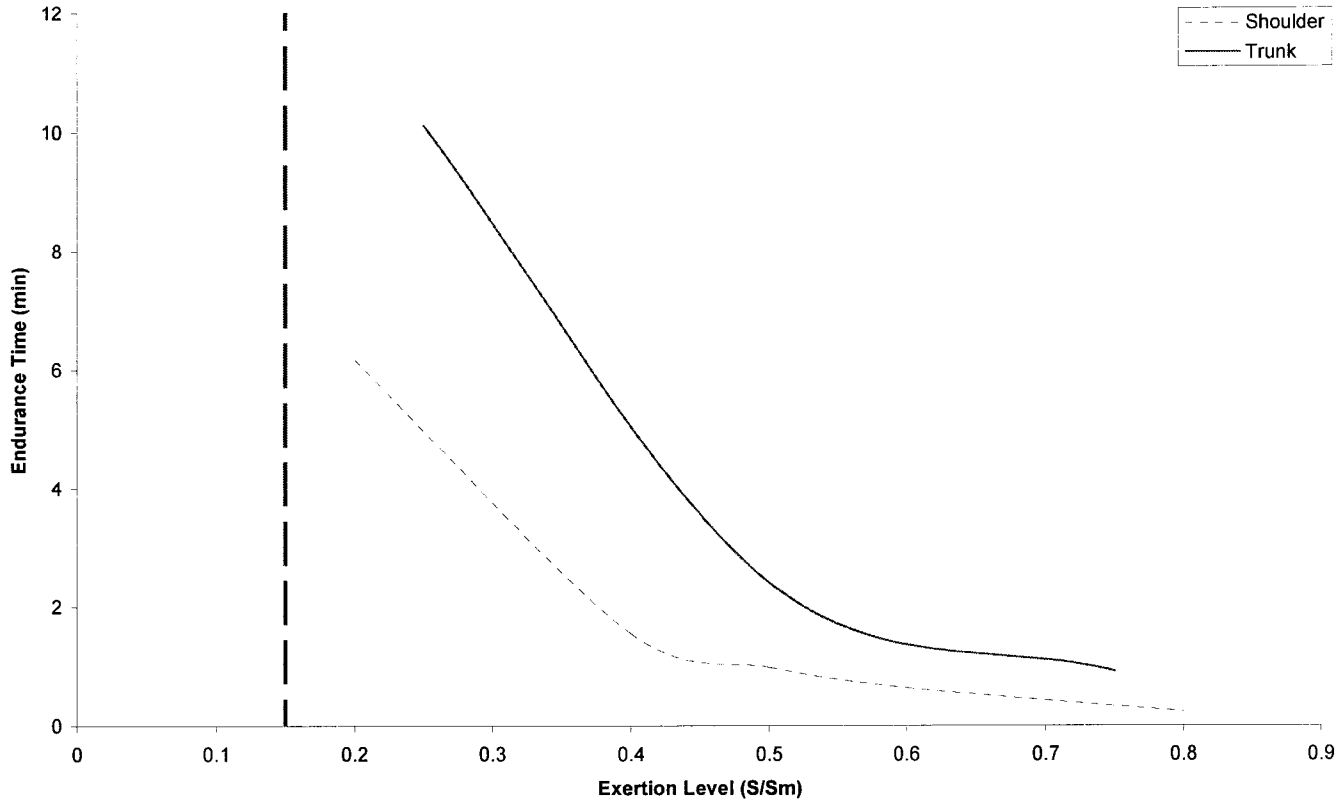

Figure 36 Static muscle endurance-exertion level relationship based on Rohmert curve

(Von Rohmert, 1960; Freund and Takala, 2001; Van Dieën et al., 1998)

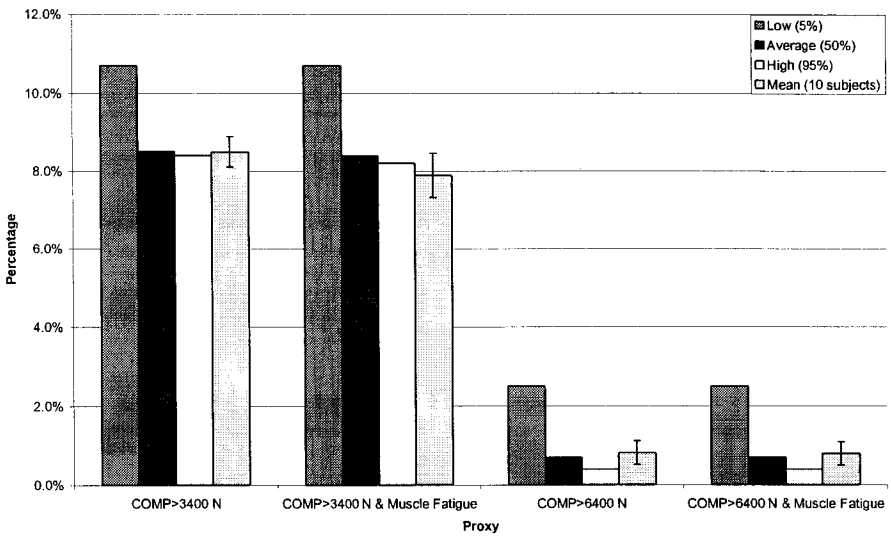

Figure 37 Frequencies of disc compression force above load tolerance and trunk muscle fatigue for 5, 50, and 95 percentiles of the hypothesized population and 10 simulated subjects during an 8-hour workday

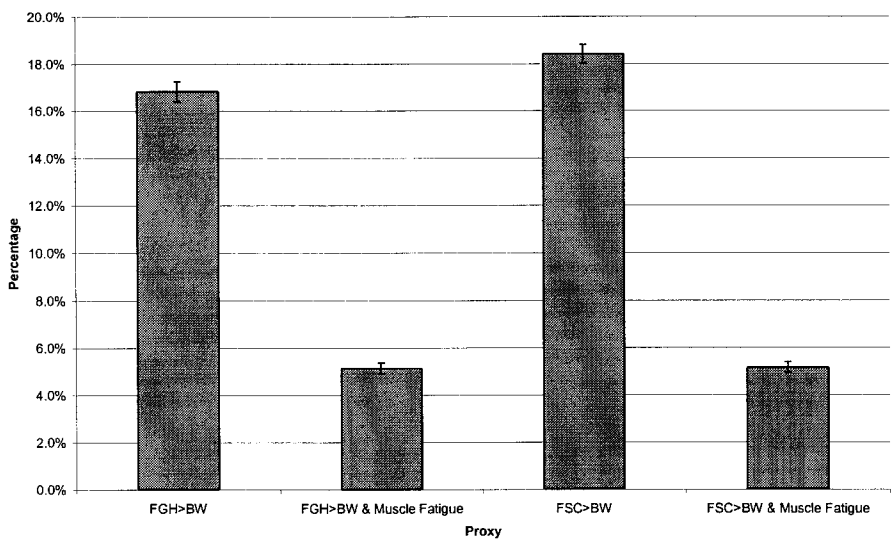

Figure 38 Frequencies of GH (glenohumeral) and SC (sternoclavicular) joint resultant forces above BW (body weight) and shoulder muscle fatigue for 10 simulated subjects during an 8-hour workday

IV. Evaluation of ergonomic intervention strategies for drywall installation

A. Introduction

The physical ergonomic exposures at the low back and shoulder during drywall installation were able to be quantified through the integration of PATH observations, Monte-Carlo simulation, and biomechanical modeling. A variety of ergonomic risk factors such as awkward postures, heavy loads in the hands, and muscle fatigue could be identified as possible contributors to musculoskeletal injuries and disorders. It is then important to implement ergonomic interventions to reduce those risks.

In general, occupational injury interventions may be grouped as engineering controls, focusing on the physical work environment; administrative interventions, addressing policies and procedures; and personal interventions, implementing worker training and personal protective equipment (Zwerling et al., 1997). Usually different interventions have been combined to be effective in the prevention and reduction of injuries, as recommended by Hsiao and Stanevich (1996). Also, many interventions are intrinsically multifactorial since the implementation of one kind of intervention may require changes in other aspects.

No matter which type of intervention is chosen, it appears that few employers will implement ergonomic interventions without sufficient evidence of their effectiveness (Zwerling et al., 1997). Conversely, it is also quite challenging to evaluate the effectiveness of any intervention before implementation. Issues related to both barriers to the adoption of risk controls and appropriate design and conduct of intervention evaluation studies arise unsurprisingly.

It is widely believed that cost and productivity are two major concerns affecting solution adoptions. Particularly, factors that govern the economics of the construction sector are incredibly complex and the costs of interventions are not only limited to economic ones, but also include other perceived costs such as the time and effort involved to acquire a risk control and to implement it at a specific worksite, and potential embarrassment (Cowley and Leggett, 2003). Other factors that influence adoption rates include architectural design, which may affect selection of materials, tools, and work methods; and union agreements, which may have an influence on the usage of ergonomic tools.

A certain ergonomic intervention will not be adopted if the expected benefits of a precaution are fewer than the costs or if the total productivity decreases for any reason. For example, based on participant reports from a stakeholder meeting, Albers et al. (2005) found that the introduction and acceptance of any ergonomic intervention to reduce musculoskeletal loading for building installation tasks was sensitive to the initial costs and its impact on productivity and workers' craft traditions. Many participants would consider ergonomic solutions based on personal or anecdotal verification, which indicated that more rigorous evaluation of the intervention effectiveness may not always be necessary for contractor trial and eventual adoption.

On the other hand, in order for injury prevention interventions to be considered effective, they must result in a reduction in ergonomic and physical stress which can be assessed by both objective and subjective measurements (Stuart and Zellers, 1996). Zwerling et al. (1997) suggested applying randomized controlled trials but recognizing large obstacles in doing so, included number and choice of units for randomization, study

group contamination, compensatory treatment in the control unit, and population turnover. Moreover, the validity of evaluations can be threatened by measurement issues, such as long latency period, multifactorial etiology of injury, and compliance with recommendations (Zwerling et al., 1997).

Ideally, since the goal of ergonomic intervention is to prevent injuries, measuring a reduction in injuries will be the most convincing demonstration of the intervention effectiveness (Zwerling et al., 1997). When there are no injury data available, the reduction of intermediate outcomes including the physical loads and muscle fatigue is selected instead, based on the hypothesized causal relationship between those variables and occupational injuries.

The objective of this study was to evaluate the efficacy of interventions to reduce the ergonomic hazards during drywall installation, utilizing the integrated biomechanical methodologies of PATH, Monte-Carlo simulation, biomechanical modeling, and the one-compartment challenge-recovery model. The principles of ergonomic interventions were firstly defined. Specific strategies were then proposed based on these and new input values for biomechanical analysis were identified for the different strategies. Values of dependent variables including the required muscle contraction forces, joint resultant forces, prevalences of muscle fatigue, and consecutive fatigue periods were obtained and compared against those under the scenario of no intervention. The biomechanical and physiological advantages of ergonomic interventions focusing on different aspects were evaluated by comparing the reductions of those dependent variable values under different intervention strategies.

B. Methods

1. Determination of dependent variables

The study integrates different methodologies of PATH, Monte-Carlo simulation, biomechanical modeling, and the one-compartment challenge-recovery model to estimate the required muscle contraction forces, joint resultant forces, and muscle fatigue parameters at the low back and shoulder during drywall installation (Figure 39). For detailed description of the methods, please refer to Chapter II and III of this dissertation (Yuan, 2006). PATH (Posture, Activity, Tools, and Handling), the observational work-sampling based approach developed by Buchholz et al. (1996), was used to quantify the percent of time that the drywall installers were conducting different activities with different body part postures. An 8-hour-workday activity series was simulated using the Monte-Carlo simulation based on activity percentages in PATH data (Step A in Figure 39). The biomechanical models of the low back developed by Marras and Granata (1995) and the shoulder model presented by Högfors et al. (1987) and tested by Karlsson and Peterson (1992) were utilized in the study to determine the musculature-associated parameters. The model input variables, including the anthropometric data, the joint angles, the external load force and position vectors, and the internal muscle parameters, were generated by Monte-Carlo simulation for the analyses of the low back and shoulder respectively (Steps B1 and B2 in Figure 39). Utilizing different optimization programs in MATLAB (The MathWorks, Natick, MA, U.S.A.), the three-dimensional static equilibrium equations were solved and the biomechanical model output variables of muscle contraction forces and joint resultant forces at the low back and shoulder for both the selected 5, 50, and 95 percentiles and 10 randomly generated subjects from the

hypothesized population (Marras et al., 2001) were obtained (Steps C1 and C2 in Figure 39). In the end, the one-compartment challenge-recovery model was applied to determine the prevalences of muscle fatigue and consecutive fatigue periods for those generated subjects (Step D in Figure 39).

2. Principles of ergonomic interventions

Basically, the principles of ergonomic interventions for drywall installation were established in an attempt to insure their effectiveness. Based on the review of previous research (Albers et al., 2005; Zwerling et al. 1997; Cowley and Leggett, 2003), several crucial requirements were summarized as:

- Low initial costs and/or the potential benefits are greater than the total costs;
- The interventions should at least not reduce productivity, which is measured by the areas of drywall sheets installed;
- Least influence on existing work methods and processes;
- Easy and feasible to implement.

Since activities of manual material handling compose the majority of drywall installation work, interventions were specifically targeted to reduce the physical ergonomic exposures during those activities. It was hypothesized that the reduction of awkward body part postures and weights in the hands would result in the decrease of muscle contraction forces and joint resultant forces, and that a better work-rest schedule would reduce muscle fatigue.

Three specific intervention strategies were proposed to reduce the ergonomic risk factors of awkward postures, heavy loads, and muscle fatigue. Similar data analysis

techniques as used to quantify ergonomic exposures (Yuan, 2006) were applied for the intervention evaluations. Particularly, since rest periods were added after activity simulation, the intervention strategy of better work-rest schedule could be examined in combination with other solutions.

1) Intervention 1 – Lifting assistance

This kind of intervention was aimed at reducing awkward trunk and shoulder postures during activity 2 (lift) and activity 3 (carry). Originally, the drywall sheets were piled horizontally on the ground. The installers always had to flex, laterally bend, and twist the trunk and raise both shoulders to lift them. The intervention was that the drywall sheets be stored vertically at an appropriate height for the installers to both lift and carry using neutral trunk and shoulder postures. Such an intervention could be accomplished through different means. The most effective one might be that the drywall sheets were stored in that ideal position when they were delivered to the construction site, so that no additional economic and labor costs were required.

Based on the biomechanical analysis of the physical loads during drywall installation, the activity frequencies in PATH under Intervention 1 would remain the same for both post- and pre- intervention conditions; however, the elimination of trunk and shoulder awkward postures were assumed to change the frequencies of body part postures associated with the activities of lift and carry (Tables 28 and 29). The distribution parameters of joint angles and external load position vectors on the Z-axis were therefore changed (Table 30). In addition, the external load position on the Y-axis

during activity 2 (lift), denoting the horizontal distance between the whole drywall sheets and the body, was halved compared to the pre-intervention condition.

2) Intervention 2 – Two-person team work

This type of intervention focused on reducing the loads in the hands during activity 2 (lift) and 3 (carry). In order not to reduce productivity, it is suggested that the two installers work as a pair and only install whole drywall sheets in the simulated 8-hour workday. The frequencies of PATH activities were changed and work cycles of installing partial sheets were deleted. The two installers were matched on anthropometric parameters, especially the subject height, for the purpose of maximizing the capacity and efficiency of team work (Lee and Lee, 2001). Frequencies of body part postures associated with different activities were assumed to remain the same as pre-intervention for the concern of least influence on existing work processes, although in reality the awkward postures may also be reduced, particularly during the lifting activity.

3) Intervention 3 – Work-rest schedule

Different from the commonly applied work-rest schedule in the construction field (15-min morning break, 30-min lunch time, and 15-min afternoon break), a new work-rest schedule reducing the maximum continuous work period from 3 hours to 1 hour was recommended. The total rest time did not change. However, workers took more frequent but shorter breaks under the proposed schedule, including five 5-min breaks in the morning, a 30-min lunch break, and another 5-min break in the afternoon. It was surmised that both muscle fatigue rates and consecutive fatigue periods could be reduced

under the new schedule whereas consecutive fatigue periods alone might vary when integrating interventions of work-rest schedule with other options.

4) Combined interventions

The above intervention strategies of lifting assistance and better work-rest schedule were combined (Intervention 1 & 3) with the goal of further reduction in ergonomic hazards for the drywall installers during the simulated 8-hour workday. Similarly, the combination of Intervention 2 (two-person team work) and Intervention 3 (work-rest schedule) were also examined.

3. Ranking of ergonomic intervention strategies

For a comprehensive evaluation of intervention efficacy and effectiveness, a set of dependent variables from the biomechanical analysis, including the means and maximums of MMCI (Maximum Muscle Contraction Intensity), the required muscle contraction forces, joint resultant forces, and muscle fatigue parameters, were considered and compared. The five intervention strategies were ranked from 1 to 5 (1 meaning the most reduction and 5 the least) according to the magnitude of their reduction in the values of those variables.

There were 10 variables for the analysis of low back, including means of MMCI, muscle forces, joint resultant forces, disc overload frequency, prevalence of muscle fatigue, consecutive fatigue periods and simultaneous muscle fatigue and disc overload, and maximums of MMCI, muscle forces, and joint resultant forces. For the shoulder, 8 of 10 variables included in the low back analysis were selected, except for the means and

maximums of MMCI which were not obtained. The scores were summed separately for each body part, and the averages for each intervention were then calculated, using the sums of the scores divided by the possible total scores of 50 and 40 for the low back and shoulder, respectively. Furthermore, the averages combining the analysis of the low back and shoulder for each intervention were obtained in order to determine the overall rankings, with a lower average denoting a higher ranking.

4. Data analysis

Subjects' anthropometric data were matched between post- and pre- interventions, since they were simulated from the same distributions using the same methods. Percent Reduction was calculated to compare the means and maximums of the required muscle contraction forces and joint resultant forces between post- and pre- interventions. Percent Reduction on prevalence of muscle fatigue, consecutive fatigue periods, and frequencies of simultaneous muscle fatigue and joint overload were also computed. The general equation for calculating Percent Reduction is:

$$Percent\ Reduction = \frac{V_{pre} - V_{post}}{V_{pre}} * 100\% \tag{4.1}$$

Where,
V_{pre} = Value of dependent variable pre- intervention
V_{post} = Value of dependent variable post- intervention

Finally, different ergonomic intervention strategies were ranked based on their roles in reducing the physical ergonomic exposures during drywall installation.

C. Results

There were no significant differences in frequencies of simulated 8-hour-workday activities between pre- and post- interventions of lifting assistance for biomechanical analysis of the low back and shoulder.

1. Low back analysis

1.1. MMCI

Table 31 lists the means and maximums (Percent Reductions) of MMCI for different subjects under different conditions of pre- and post- intervention. The means of MMCI for 5, 50, and 95 percentiles of the hypothesized population under the intervention of lifting assistance were 58.1, 38.6, and 32.4 N/cm^2 respectively, which indicated 17.5%, 11.5%, and 9.7% reductions respectively compared to those values pre- intervention (Figure 40). The maximum values of MMCI decreased too and the Percent Reductions were greater than those for the mean values (Figure 41). On the other hand, the average of MMCI means for 10 simulated subjects was 37.3 N/cm^2, 9% lower than the value of 41.0 N/cm^2 that was under no intervention (Figure 40). The average of MMCI maximum values decreased 26.9% after Intervention 1 (Figure 41).

With the intervention of two-person team work, the means of MMCI for 5, 50, and 95 percentiles of the hypothesized population were 64.2, 41.9, and 34.2 N/cm^2 respectively, showing 8.9%, 3.9%, and 4.7% reductions respectively against pre-intervention (Figure 40). Their maximum values dropped more than for Intervention 1 (Figure 41).

The MMCI for 10 simulated subjects under Intervention 2 was 39.0 N/cm^2 on average, which was 5% lower than pre- intervention. Nonetheless, the average of MMCI maximum values decreased 34.3% and the reduction was greater than for Intervention 1 (Figure 41).

1.2. Required muscle contraction forces

Means of the required muscle contraction forces for the major trunk muscle groups (Table 32), including LEL, LER, LDL, LDR, IOL, and IOR, were all greatly reduced under Intervention 1 for 5, 50, and 95 percentiles of the hypothesized population, respectively (Figures 42, 43 and 44). The maximum muscle contraction forces were lowered more apparently (Table 33, and Figures 45, 46, and 47). For the comparison of 10 simulated subjects, there were at least 10% reductions for those trunk muscles (Figure 48). Moreover, the averages of maximum muscle forces for those six muscle groups were all reduced more than 20% after Intervention 1 (Figure 49).

The required muscle contraction forces were reduced under Intervention 2 for different anthropometries in four major trunk muscle groups, including LEL, LER, LDL, and LDR (Figures 42, 43 and 44). While the reductions in mean values were less than those under Intervention 1, the maximum muscle forces decreased more notably (Figure 45, 46, and 47). Similar patterns occurred in the reductions of muscle contraction forces for 10 simulated subjects (Figures 48 and 49).

1.3. Joint resultant forces

Joint resultant forces of disc compression, lateral shear and anterior-posterior shear (Tables 34 and 35) were all reduced on average for 5, 50, and 95 percentiles of the hypothesized population after Intervention 1 (Figures 50, 51 and 52). The highest values dropped dramatically as well (Figures 53, 54, and 55). The averages of joint resultant force means for 10 simulated subjects were all lowered more or less after Intervention 1 (Figure 56). The maximum joint resultant forces showed a greater Percent Reduction (Figure 57). As a result, disc compression force never exceeded 6400 N.

Under Intervention 2, reductions in the means of joint resultant forces of disc compression and anterior-posterior shear were found for 5, 50, and 95 percentiles of the hypothesized population (Figures 50, 51 and 52). In terms of the lateral shear force, there were reductions for both the 5 and 95 percentiles, but not for the 50 percentile of the hypothesized population. The maximum values of joint resultant forces decreased more visibly than those under Intervention 1 (Figures 53, 54, and 55).

Both the mean and maximum values of joint resultant forces for 10 simulated subjects were reduced on average under Intervention 2, except for the lateral shear force means. However, the reductions were more notable than those under Intervention 1, especially for the maximum values of disc compression and anterior-posterior shear forces (Figure 57).

2. Shoulder analysis

2.1. Required muscle contraction forces

Fairly large decreases were found in the averages of muscle contraction force means for a number of major shoulder muscles after Intervention 1, such as DE(A), DE(M), IN(S), IN(I), Sub(S), Sub(M), Sub(I), Sup, TMi, and COL (Figure 58). The maximum muscle contraction forces for most shoulder muscles included in the study were reduced for the 10 simulated subjects on average (Figure 59).

Under Intervention 2, muscle contraction force means of LD(S), LD(I), PMa(S), PMa(C), DE(A), DE(P), CO, TMi, and COL decreased tremendously on average, whereas muscle force means of Sup, TMa, BB(L), and BB(S) increased surprisingly (Figure 58). It was surmised that installing only whole sheets under Intervention 2 might have increased the general physical requirements on those muscles. On the other hand, there were great decreases in the maximum contraction forces for almost every shoulder muscle, and the reductions were more remarkable than those under Intervention 1 (Table 36 and Figure 59).

2.2. Joint resultant forces

In contrast to joint resultant forces at the low back, both the averages of joint resultant force means at GH and SC showed reductions after Intervention 1, as well as the maximum values (Figures 60 and 61). There were about 6% and 3% reductions in joint overload rate at GH and SC respectively, when using the body weight as the criterion.

Intervention 2 brought reductions in both the mean and maximum values of joint resultant forces at GH and SC, and the reductions in the maximum values were much

more greater compared to Intervention 1 (Figures 60 and 61). Interestingly, joint overload frequencies at both GH and SC increased about 20% (Table 37), possibly because that more material handling activities were involved under Intervention 2 based on the assumptions.

3. Muscle fatigue

3.1. Prevalences of muscle fatigue

The prevalences of trunk muscle fatigue showed decreases for both 5, 50, and 95 percentiles of the hypothesized population and 10 simulated subjects after different types of interventions, except for 5 percentile under Intervention 2 and Intervention 2 & 3 (Table 38 and Figure 62). The reduction of the average frequency for 10 simulated subjects under Intervention 1 was very remarkable (47.4%) and became even more productive (53%) when combined with Intervention 3 (Figure 62).

Since the maximum disc compression forces never exceeded 6400 N under Interventions 1 and 2, the prevalences of simultaneous muscle fatigue and disc compression forces above the load tolerances (especially for 6400 N) for different anthropometries were reduced (Figures 63, 64, 65, and 66). However, frequencies of simultaneous muscle fatigue and disc compression forces above 3400 N for both 50 and 95 percentiles of the hypothesized population were increased under Intervention 2 (Table 39).

All five different interventions brought reductions in shoulder muscle fatigue frequencies, where Intervention 1 and Intervention 1 & 3 tended to be relatively more

effective (Figure 67). Frequencies of simultaneous muscle fatigue and disc overload at GH and SC were also reduced (Figure 68 and Table 40).

3.2. Consecutive fatigue periods

Intervention 1 and Intervention 1 & 3 produced relatively greater reductions in the total number and average durations of trunk muscle consecutive fatigue periods for 10 simulated subjects (Table 41). Reductions were also evident for 5, 50, and 95 percentiles of the hypothesized population under those interventions. On the other hand, the number and average durations of trunk muscle consecutive fatigue periods varied under Intervention 2 and Intervention 2 & 3. There were no such periods for both 50 and 95 percentiles of the hypothesized population; however, the number increased for the 5 percentile (Figure 69). The number of periods decreased for 10 simulated subjects, but such periods lasted longer on average compared to pre-intervention (Figure 70).

There were no consecutive fatigue periods for the shoulder muscles found under any intervention, which once again indicated the intermittence of shoulder muscle fatigue.

4. Ranking of ergonomic interventions

It turned out that the combination of Intervention 1 & 3 was the most efficacious intervention strategy to reduce the physical loads on the low back and shoulder for drywall installers, followed closely by Intervention 1, Intervention 2 & 3, and Intervention 2 (Table 42). Intervention 3 helped to reduce muscle fatigue, although it was much less beneficial in reducing the muscle forces and joint resultant forces. Intervention

1 & 3 and Intervention 2 & 3 were rated as equally efficacious when comparing the reductions at the low back only. For the shoulder, Intervention 1 and Intervention 1 &3 showed more efficacy than Intervention 2 and Intervention 2 & 3.

D. Discussion

Generally, all of the proposed interventions produced reductions in the physical loads on the low back and shoulder during drywall installation. The interventions of lifting assistance and two-person team work helped to reduce the required muscle contraction forces and joint resultant forces, and the intervention of work-rest schedule was able to mitigate muscle fatigue. Particularly when the intervention of work-rest schedule was combined with either of the first two interventions, it contributed to minimizing muscle fatigue.

In fact, Intervention 1 was more efficacious in reducing the mean values of those variables being compared, whereas Intervention 2 produced more reductions in the maximum values. If it sounds reasonable that the mean values could be linked to the cumulative physical burden and the maximum ones, on the other hand, could be associated with the acute load, the two interventions could be then prioritized differently. For example, when cumulative trauma disorders are frequent, Intervention 1 should be prioritized and contrarily in a workforce where acute injuries frequently occur, it is highly recommended to implement Intervention 2.

Since Intervention 3 dealt with work-rest schedule only and did not involve changes of work methods, postures, and tools, it was attached to the first two interventions during the analysis. Even for the least investment of any intervention, Intervention 3 alone proved to be effective in reducing muscle fatigue (Figures 62 and 67).

It should be recognized that the principles of ergonomic interventions proposed in this study were solely based on the summary from literature review (Zwerling et al. 1997;

Cowley and Leggett, 2003). The basic hypothesis for any specific intervention strategy was to produce reductions in ergonomic hazards with the relatively little change in different parameters, such as cost, productivity, work methods and processes, etc. It is very difficult to justify the change especially in productivity; on the other hand, it may involve changes in multiple aspects in reality, which was simplified in this study.

The assumption of least change in PATH data under different interventions was another example of satisfying the proposed guidelines for ergonomic interventions. In reality, the frequencies of activities and body part postures after a certain intervention may change a lot, instead of assumed to remain the same for the comparisons of dependent variables between post- and pre- intervention. For example, under the intervention of two-person team work, the team members may reduce the awkward trunk postures when lifting the drywall sheets from the two long sides, instead of using the same lifting technique that was applied for one worker before intervention (Yuan, 2006). The results showed that the physical loads and muscle fatigue were reduced when keeping the frequencies of body part postures associated with different activities unchanged under Intervention 2. It was thus plausible to expect even greater reductions to be realized should the team members have changed the lifting patterns.

The five intervention strategies were ranked semi-quantitatively based on the Percent Reductions of all dependent variables. The correlations among different variables were neglected since it was not examined. Considering that all dependent variables were closely related to each other, it could be imagined that the reduction in one variable might automatically result in the reduction in all variables though the Percent Reductions were different. Furthermore, the ranking of 1 to 5 for a specific variable (for example, muscle

contraction force means) was based on the consideration of all components included in that variable. Thus, the ranking was targeted to compare the relative reductions, rather than the absolute amount.

This study utilized intermediate outcomes such as muscle forces, joint resultant forces, muscle fatigue frequencies, and consecutive fatigue periods to evaluate the efficacy of the intervention strategies. Different from traditional studies which are either focused on the experimental trials of efficacy evaluation, or targeted at the actual field application of effectiveness assessment, or both, this research examined both the efficacy and "pseudo" effectiveness in a relatively easy way with the help of computer simulations. Of course, it should be noted that the single 8-hour workday simulated in this study was quite short in terms of demonstrating effectiveness in the long term. However, the entire evaluation techniques follow positively in the direction suggested by Zwerling et al. (1997), where simple quasi-experimental design, randomized control, and measurement of intermediate outcomes were all involved more or less.

These analyses evaluated the influences that the proposed interventions would have on intermediate outcomes, rather than investigating the practical implementations of those interventions. For the sake of more feasible field applications, it is necessary to clarify potential barriers to intervention adoption.

In terms of Intervention 1, it may be an issue as to whether or not it is the drywall delivery company's responsibility to deliver the drywall sheets into the ideal position for the installers to handle manually. While the contractors could also possibly purchase some mechanical devices for lifting assistance, however, it is usually difficult to convince

them since on one hand, there is no legal requirement and on the other hand, the contractors do not believe they benefit from this kind of investment.

Also, how to actually store the drywall sheets if they are suggested to stay in the ideal position is another question. Since the drywall sheets commonly used in the commercial construction industry are very heavy and bulky, it is undoubtedly hard to have them stored vertically without any mechanical assistance. Safety concerns may also arise if they are not firmly secured.

As naturally assumed and frequently observed in the field, drywall installation is usually conducted by one worker. It is not uncommon that the drywall installers will be reluctant to make changes in both their minds and the way they actually do during the intervention of two-person team work. Moreover, the manual handling guidelines in Australia typically recommend that individuals involved in team lifts should be of similar standing height (NOHSC, 1990; DETIR, 2000). Past research also indicated that matching team member height may be important for reducing cumulative loading during repetitive lifting (Dennis and Barrett, 2003). This would be difficult for contractors who either have smaller workforces or have high turnover rates.

Using the areas of drywall sheets installed as the criteria for measuring the productivity of drywall installation, workers in the team were required to install the whole drywall sheet exclusively, in order to cover approximately twice the area as pre-intervention. Eventually other workers would have to deal with partial sheets, because many building structures require smaller pieces of drywall to be installed. The intervention of team work will inevitably cause organizational change, where various factors, especially work experience, should also be considered during the team selection.

The intervention of work-rest schedule promises to attenuate muscle fatigue by reducing the maximum consecutive work period from 3 hours to 1 hour. More frequent but shorter rest periods were taken and muscle fatigue was indeed predicted to be reduced. Nonetheless, workers' morning break (usually for breakfast and coffee) was truncated. Furthermore, it is unrealistic to follow such a schedule especially under the push of a project deadline when even the regular rest schedule will be threatened.

It was assumed in the study that all of the simulated subjects were able to finish the simulated 8-hour-workday activities without any musculoskeletal injuries and/or disorders. In reality, drywall installers might have been injured at the moment of peak muscle forces and/or joint resultant forces and thus have had to change their behaviors, including postures and work methods, after that moment or even to stop work. If that commonly occurs in the field, those interventions which are beneficial in reducing the maximum forces, such as the intervention of two-person team work examined in this study, should be prioritized.

There also exist other kinds of intervention, of which manufacturing smaller and lighter drywall sheets may be the crucial one. Lappalainen et al. (1998) found that lighter and narrower plasterboard had a significant influence on the physical loads and risks of musculoskeletal diseases among the installers. However, they argued that full health benefits could only be achieved if workers were also taught appropriate handling techniques. Further, reducing the size of drywall sheets may depend on the needs of both economic markets and architectural structures.

In summary, significant intervention strategies could be implemented to both reduce the physical ergonomic hazards and promote workers' safety and health during

drywall installation. Cost-effective approaches to evaluating ergonomic interventions are warranted to firstly address their efficacy and effectiveness and then to maximize the adoption of evaluated controls.

E. Conclusions

The efficacy of ergonomic interventions for drywall installation were evaluated integrating different methodologies, a work-sampling based approach (PATH), computer-aided simulation (Monte-Carlo simulation), biomechanical modeling, and a one-compartment challenge-recovery model. Intermediate outcomes such as muscle contraction forces, joint resultant forces, prevalences of muscle fatigue, and muscle consecutive fatigue periods, were selected as variables to delineate the biomechanical advantages of different intervention strategies. Among those proposed ergonomic interventions including lifting assistance, two-person team work, and optimal work-rest schedule, the combination of lifting assistance and optimal work-rest schedule proved to be the most efficacious in reducing both the physical hazards and muscle fatigue on the low back and shoulder for drywall installers.

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Table 28

Idealized frequencies of PATH trunk posture data associated with different activities under Intervention 1 (lifting assistance)

Activity	Amount	Neutral	Mild Flexion	Severe Flexion	Bend & Twist	Bend & Twist & Flexion	Activity Percent	Neutral	Mild Flexion	Severe Flexion	Bend & Twist	Bend & Twist & Flexion
1 (cut/measure)	25	12	4	3	1	5	19.8%	9.5%	3.2%	2.4%	0.8%	4.0%
2 (lift)	8	**8**	**0**	**0**	**0**	**0**	**6.4%**	**6.4%**	**0.0%**	**0.0%**	**0.0%**	**0.0%**
3 (carry)	16	**16**	**0**	**0**	**0**	**0**	**12.7%**	**12.7%**	**0.0%**	**0.0%**	**0.0%**	**0.0%**
4 (hold/place)	6	2	1	2	1	0	4.8%	1.6%	0.8%	1.6%	0.8%	0.0%
5 (screw)	20	7	5	3	3	2	15.9%	5.6%	4.0%	2.4%	2.4%	1.6%
6 (in between)	24	22	2	0	0	0	19.1%	17.5%	1.6%	0.0%	0.0%	0.0%
7 (other)	27	13	2	8	0	4	21.4%	10.3%	1.6%	6.4%	0.0%	3.2%
Total	126	71	16	18	5	16	100.0%	56.4%	12.7%	14.3%	4.0%	12.7%

Note: bold means changes from pre- intervention.

Table 29

Idealized frequencies of PATH shoulder posture data associated with different activities under Intervention 1 (lifting assistance)

Activity	Amount	2 down	1 up	2 up	Percent	2 down	1 up	2 up
1 (cut/measure)	25	13	4	8	19.8%	10.3%	3.2%	6.4%
2 (lift)	8	**8**	**0**	**0**	**6.4%**	**6.4%**	**0.0%**	**0.0%**
3 (carry)	16	**16**	**0**	**0**	**12.7%**	**12.7%**	**0.0%**	**0.0%**
4 (hold/place)	6	3	0	3	4.8%	2.4%	0.0%	2.4%
5 (screw)	20	9	6	5	15.9%	7.1%	4.8%	4.0%
6 (in between)	24	23	0	1	19.1%	18.3%	0.0%	0.8%
7 (other)	27	18	2	7	21.4%	14.3%	1.6%	5.6%
Total	126	80	16	30	100.0%	63.5%	12.7%	23.8%

Note: bold means changes from pre- intervention.

Table 30

Distribution parameters for trunk and shoulder postures and external load position vectors on the Z-axis under Intervention 1 (lifting assistance)

Variable		Distribution Type	Distribution Parameter	
			Mean/GM (degree)	SD/GSD (degree)
Trunk Posture	i	Lognormal	2.8	0.7
	j	Normal	2.1	9.6
	k	Normal	-2.5	11.4
i-shoulder		Lognormal	3.7	0.6
ez		Normal	0.16	0.35

Note:

i – Flexion; j – Lateral bend; k – Twist; GM – Geometric Mean; SD – Standard Deviation; GSD – Geometric Standard Deviation; ez – External load position vectors on the Z-axis.

Table 31

MMCI (Maximum Muscle Contraction Intensity) means and maximums (Percent Reductions) for different anthropometries under different situations

Mean	Pre-	Post-1	Post-2
Low (5%)	70.5	58.1 (17.5%)	64.2 (8.9%)
Average (50%)	43.6	38.6 (11.5%)	41.9 (3.9%)
High (95%)	35.9	32.4 (9.7%)	34.2 (4.7%)
Mean (10 subjects)	41.0	37.3 (9.0%)	39.0 (5.0%)
Max	**Pre-**	**Post-1**	**Post-2**
Low (5%)	367.8	274.8 (25.3%)	228.7 (37.8%)
Average (50%)	194.8	149.3 (23.4%)	129.2 (33.7%)
High (95%)	137.1	108.3 (21.0%)	97.0 (29.2%)
Mean (10 subjects)	178.2	130.3 (26.9%)	117.1 (34.3%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work.

Table 32

Means (Percent Reductions) of the required trunk muscle contraction forces for different anthropometries under different situations

Anthropometry	Situation	LEL	LER	LDL	LDR	EOL	EOR	IOL	IOR	RAL	RAR
5%	Pre-	1331.9	1386.6	865.3	854.4	0.5	22.4	113.7	90.3	0.0	0.0
	Post-1	1094.1 (17.9%)	1137.1 (18.0%)	701.8 (18.9%)	696.5 (18.5%)	0.5 (-2.1%)	18.0 (19.4%)	74.0 (34.9%)	79.4 (12.1%)	0.1 (-222.7%)	0.0 (-13.9%)
	Post-2	1203.6 (9.6%)	1261.8 (9.0%)	782.5 (9.6%)	778.5 (8.9%)	0.2 (68.1%)	18.5 (17.4%)	96.2 (15.4%)	87.4 (3.2%)	0.0 (16.2%)	0.1 (-168.2%)
50%	Pre-	1074.5	1082.8	648.1	665.2	0.5	5.1	91.2	93.6	0.0	0.2
	Post-1	944.4 (12.1%)	954.8 (11.8%)	549.2 (15.3%)	567.1 (14.7%)	0.5 (2.6%)	3.2 (38.0%)	67.5 (26.0%)	88.2 (5.8%)	0.0 (-72.7%)	0.0 (82.0%)
	Post-2	1029.3 (4.2%)	1050.0 (3.0%)	622.1 (4.0%)	640.5 (3.7%)	0.5 (-3.7%)	4.2 (18.1%)	87.4 (4.2%)	96.6 (-3.2%)	0.0 (52.8%)	0.1 (44.3%)
95%	Pre-	984.7	984.0	619.5	650.9	1.9	1.6	88.6	103.2	0.0	0.1
	Post-1	877.2 (10.9%)	879.7 (10.6%)	525.2 (15.2%)	551.0 (15.4%)	0.5 (74.1%)	0.7 (58.8%)	69.4 (21.6%)	97.6 (5.4%)	0.0 (-428.1%)	0.0 (95.8%)
	Post-2	929.7 (5.6%)	942.5 (4.2%)	589.3 (4.9%)	621.2 (4.6%)	3.2 (-72.4%)	1.4 (15.2%)	81.8 (7.6%)	103.8 (-0.6%)	0.0 (34.7%)	0.1 (32.2%)
Mean (10 Subjects)	Pre-	989.7	1009.7	591.0	603.8	3.6	12.1	93.5	90.7	0.1	0.2
	Post-1	877.4 (11.4%)	894.2 (11.4%)	533.8 (9.7%)	546.2 (9.5%)	2.2 (37.4%)	9.6 (20.5%)	64.9 (30.6%)	81.5 (10.1%)	0.1 (14.4%)	0.1 (48.8%)
	Post-2	927.7 (6.3%)	952.4 (5.7%)	583.5 (1.3%)	593.8 (1.7%)	11.7 (-225.6%)	5.2 (56.9%)	74.3 (20.5%)	83.5 (8.0%)	0.0 (77.1%)	0.1 (51.1%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work; LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; LDL – Left Latissimus Dorsi; LDR – Right Latissimus Dorsi; EOL – Left External Oblique; EOR – Right External Oblique; IOL – Left Internal Oblique; IOR – Right Internal Oblique; RAL – Left Rectus Abdominus; RAR – Right Rectus Abdominus.

Table 33

Maximums (Percent Reductions) of the required trunk muscle contraction forces for different anthropometries under different situations

Anthropometry	Situation	LEL	LER	LDL	LDR	EOL	EOR	IOL	IOR	RAL	RAR
5%	Pre-	7119.9	7363.6	4891.6	5511.3	260.9	1654.5	2293.6	1884.2	26.3	39.7
	Post-1	5319.6 (25.3%)	5501.7 (25.3%)	3656.6 (25.2%)	4268.1 (22.6%)	292.8 (-12.2%)	1181.8 (28.6%)	1087.9 (52.6%)	958.8 (49.1%)	73.4 (-179.5%)	55.1 (-38.8%)
	Post-2	4427.2 (37.8%)	4578.7 (37.8%)	3041.9 (37.8%)	3479.0 (36.9%)	218.4 (16.3%)	1029.3 (37.8%)	1630.1 (28.9%)	1076.3 (42.9%)	37.0 (-41.0%)	147.5 (-271.7%)
50%	Pre-	5152.3	5135.9	3713.2	4048.6	249.4	767.0	1711.4	1403.0	51.8	143.9
	Post-1	3948.9 (23.4%)	3936.3 (23.4%)	2845.8 (23.4%)	3161.4 (21.9%)	276.1 (-10.7%)	532.6 (30.6%)	905.0 (47.1%)	896.9 (36.1%)	104.0 (-100.6%)	41.3 (71.3%)
	Post-2	3417.3 (33.7%)	3403.7 (33.7%)	2465.7 (33.6%)	2631.1 (35.0%)	277.2 (-11.2%)	483.1 (37.0%)	1369.4 (20.0%)	1068.8 (23.8%)	23.7 (54.3%)	81.6 (43.3%)
95%	Pre-	4179.7	4109.8	3500.8	3558.5	648.6	666.3	1541.4	1326.2	23.8	161.0
	Post-1	3301.7 (21.0%)	3246.5 (21.0%)	2773.2 (20.8%)	2879.9 (19.1%)	309.2 (52.3%)	265.7 (60.1%)	863.5 (44.0%)	858.3 (35.3%)	95.7 (-302.1%)	13.8 (91.4%)
	Post-2	2957.2 (29.2%)	2889.8 (29.7%)	2481.1 (29.1%)	2506.6 (29.6%)	452.7 (30.2%)	311.5 (53.3%)	1275.8 (17.2%)	843.9 (36.4%)	26.3 (-10.6%)	165.5 (-2.8%)
Mean (10 Subjects)	Pre-	4785.9	4686.4	3363.4	3727.8	522.2	969.0	1829.1	1344.8	91.4	140.9
	Post-1	3458.4 (27.7%)	3446.1 (26.5%)	2570.3 (23.6%)	2821.7 (24.3%)	468.6 (10.3%)	716.8 (26.0%)	992.0 (45.8%)	991.6 (26.3%)	69.6 (23.9%)	71.3 (49.4%)
	Post-2	3119.4 (34.8%)	3104.2 (33.8%)	2428.6 (27.8%)	2486.0 (33.3%)	558.1 (-6.9%)	509.3 (47.4%)	1113.7 (39.1%)	965.2 (28.2%)	40.3 (56.0%)	85.4 (39.4%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work; LEL – Left Lumbar Erector Spinae; LER – Right Lumbar Erector Spinae; LDL – Left Latissimus Dorsi; LDR – Right Latissimus Dorsi; EOL – Left External Oblique; EOR – Right External Oblique; IOL – Left Internal Oblique; IOR – Right Internal Oblique; RAL – Left Rectus Abdominus; RAR – Right Rectus Abdominus.

Table 34

Means (Percent Reductions) of joint resultant forces at L4/L5 for different anthropometries under different situations

Anthropometry	Situation	COMP	ABS(LS)	ABS(APS)
5%	Pre-	1606.4	386.8	571.8
	Post-1	1368.0 (14.8%)	327.7 (15.3%)	477.3 (16.5%)
	Post-2	1476.1 (8.1%)	358.8 (7.2%)	521.3 (8.8%)
50%	Pre-	1729.0	319.6	475.7
	Post-1	1570.2 (9.2%)	295.1 (7.7%)	413.1 (13.1%)
	Post-2	1680.9 (2.8%)	322.7 (-1.0%)	449.9 (5.4%)
95%	Pre-	1925.0	307.7	503.6
	Post-1	1771.4 (8.0%)	290.0 (5.8%)	440.6 (12.5%)
	Post-2	1856.7 (3.5%)	307.0 (0.2%)	475.7 (5.6%)
Mean (10 Subjects)	Pre-	1724.9	291.1	487.5
	Post-1	1683.4 (2.4%)	270.5 (7.1%)	388.7 (20.3%)
	Post-2	1621.6 (6.0%)	297.4 (-2.2%)	380.4 (22.0%)

Note:

Post-1 – Lifting assistance; Post-2 – Two-person team work; COMP – Disc compression force; ABS(LS) – Absolute value of lateral shear force; ABS(APS) – Absolute value of anterior-posterior shear force.

Table 35

Maximums (Percent Reductions) of joint resultant forces at L4/L5 for different anthropometries under different situations

Anthropometry	Situation	COMP	ABS(LS)	ABS(APS)
5%	Pre-	7430.6	3273.5	3919.6
	Post-1	5452.2 (26.6%)	2251.5 (31.2%)	2768.7 (29.4%)
	Post-2	4628.3 (37.7%)	2170.8 (33.7%)	2374.8 (39.4%)
50%	Pre-	6929.6	2734.5	2866.6
	Post-1	5342.3 (22.9%)	1637.5 (40.1%)	2129.5 (25.7%)
	Post-2	4861.1 (29.9%)	1823.0 (33.3%)	1914.2 (33.2%)
95%	Pre-	6829.2	2301.4	2787.5
	Post-1	5381.2 (21.2%)	1425.6 (38.1%)	2038.0 (26.9%)
	Post-2	4958.9 (27.4%)	1526.1 (33.7%)	1844.5 (33.8%)
Mean (10 Subjects)	Pre-	6899.9	2385.7	3038.0
	Post-1	5877.4 (14.8%)	1712.7 (28.2%)	2117.3 (30.3%)
	Post-2	4602.1 (33.3%)	1720.5 (27.9%)	1622.8 (46.6%)

Note:

Post-1 – Lifting assistance; Post-2 – Two-person team work; COMP – Disc compression force; ABS(LS) – Absolute value of lateral shear force; ABS(APS) – Absolute value of anterior-posterior shear force.

Table 36

Means and maximums (Percent Reductions) of the required shoulder muscle contraction forces under different situations

Parameter	Situation	LD(S)	LD(I)	PMa(S)	PMa(C)	DE(A)	DE(M)	DE(P)	CO	IN(S)	IN(I)
Mean	Pre-	6.9	6.5	27.4	10.8	1.0	1.7	6.1	4.0	0.3	0.8
	Post-1	6.5 (5.0%)	6.1 (6.1%)	27.5 (-0.6%)	10.7 (1.1%)	0.6 (42.5%)	1.2 (27.9%)	5.9 (4.1%)	3.7 (8.4%)	0.2 (46.1%)	0.5 (42.1%)
	Post-2	3.5 (48.4%)	3.3 (49.7%)	17.8 (35.0%)	6.4 (40.4%)	0.8 (16.4%)	1.8 (-7.7%)	5.1 (17.4%)	1.9 (51.4%)	0.2 (26.2%)	0.9 (-8.5%)
Max	Pre-	207.6	245.4	341.1	233.8	351.9	428.6	432.3	197.2	234.5	330.8
	Post-1	201.0 (3.2%)	239.9 (2.3%)	333.8 (2.2%)	217.1 (7.2%)	318.9 (9.4%)	423.0 (1.3%)	402.5 (6.9%)	171.8 (12.9%)	174.2 (25.7%)	316.5 (4.3%)
	Post-2	72.5 (65.1%)	109.4 (55.4%)	294.0 (13.8%)	204.6 (12.5%)	311.0 (11.6%)	421.8 (1.6%)	386.5 (10.6%)	143.2 (27.4%)	175.5 (25.2%)	316.0 (4.5%)
Parameter	**Situation**	**Sub(S)**	**Sub(M)**	**Sub(I)**	**Sup**	**TMa**	**TMi**	**BB(L)**	**BB(S)**	**TRI**	**COL**
Mean	Pre-	25.2	23.6	30.5	61.5	52.4	6.0	16.8	77.2	51.5	233.6
	Post-1	23.1 (8.3%)	22.0 (6.6%)	28.5 (6.6%)	58.7 (4.4%)	51.7 (1.3%)	5.1 (15.3%)	16.3 (2.8%)	75.8 (1.7%)	51.1 (0.9%)	208.5 (10.8%)
	Post-2	24.6 (2.6%)	22.9 (2.8%)	30.5 (0.2%)	66.4 (-8.0%)	65.7 (-25.4%)	5.5 (8.8%)	18.3 (-9.1%)	89.3 (-15.7%)	51.3 (0.4%)	191.7 (18.0%)
Max	Pre-	294.3	273.8	347.2	289.7	387.2	191.7	226.2	209.2	666.8	3257.5
	Post-1	264.8 (10.0%)	243.1 (11.2%)	315.6 (9.1%)	285.2 (1.6%)	348.4 (10.0%)	171.3 (10.6%)	195.1 (13.8%)	175.4 (16.1%)	645.0 (3.3%)	3142.5 (3.5%)
	Post-2	260.6 (11.5%)	238.9 (12.8%)	308.9 (11.0%)	268.1 (7.5%)	345.5 (10.8%)	169.9 (11.4%)	191.4 (15.4%)	170.3 (18.6%)	643.8 (3.4%)	1852.5 (43.1%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work; LD(S) – Latissimus Dorsi (Superior); LD(I) – Latissimus Dorsi (Inferior); PMa(S) – Pectoralis Major (Sternum); PMa(C) – Pectoralis Major (Clavicle); DE(A) – Deltoideus (Anterior); DE(M) – Deltoideus (Medial); DE(P) – Deltoideus (Posterior); CO – Coracobrachialis; IN(S) – Infraspinatus (Superior); IN(I) – Infraspinatus (Inferior); Sub(S) – Subscapularis (Superior); Sub(M) – Subscapularis (Middle); Sub(I) – Subscapularis (Inferior); Sup – Supraspinatus; TMa – Teres Major; TMi – Teres Minor; BB(L) – Biceps Brachii (Long); BB(S) – Biceps Brachii (Short); TRI – Triceps; COL – Coracohumeral Ligament.

Table 37

Means and maximums (Percent Reductions) of joint resultant forces at GH (glenohumeral) and SC (sternoclavicular) under different situations

Parameter	Situation	FGH	FSC
Mean	Pre-	480.9	574.4
	Post-1	441.4 (8.2%)	547.1 (4.8%)
	Post-2	451.6 (6.1%)	542.1 (5.6%)
Max	Pre-	4158.3	4237.5
	Post-1	3902.2 (6.2%)	4042.1 (4.6%)
	Post-2	2343.7 (43.6%)	2502.4 (40.9%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work.

Table 38

Prevalences (Percent Reductions) of trunk and shoulder muscle fatigue for different anthropometries under different situations

Body Part	Anthropometry	Pre-	Post-3	Post-1	Post-1&3	Post-2	Post-2&3
Trunk	Low (5%)	57.3%	55.9% (2.4%)	42.0% (26.7%)	40.1% (30.0%)	73.5% (-28.3%)	70.0% (-22.2%)
	Average (50%)	40.4%	37.3% (7.7%)	18.4% (54.5%)	15.7% (61.1%)	34.8% (13.9%)	29.9% (26.0%)
	High (95%)	27.9%	23.8% (14.7%)	6.0% (78.5%)	4.4% (84.2%)	15.7% (43.7%)	13.6% (51.3%)
	Mean (10 subjects)	33.5%	32.3% (3.6%)	17.6% (47.4%)	15.7% (53.0%)	27.4 (18.1%)	25.7% (23.1%)
Shoulder	Mean (10 subjects)	5.1%	4.8% (5.7%)	3.6% (29.1%)	3.4% (33.8%)	4.2% (17.2%)	4.0% (21.3%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work; Post-3 – Work-rest schedule.

Table 39

Prevalences (Percent Reductions) of disc compression forces above load tolerances and simultaneous trunk muscle fatigue and disc compression forces above load tolerances for different anthropometries under different situations

Anthropometry	Situation	COMP > 3400 N	COMP > 3400 N & Muscle Fatigue	COMP > 6400 N	COMP > 6400 N & Muscle Fatigue
5%	Pre-	10.7%	10.7%	2.5%	2.5%
	Post-1	6.6% (38.4%)	6.4% (40.0%)	0.0% (100.0%)	0.0% (100.0%)
	Post-2	9.3% (13.1%)	9.1% (15.0%)	0.0% (100.0%)	0.0% (100.0%)
50%	Pre-	8.5%	8.4%	0.7%	0.7%
	Post-1	4.2% (50.9%)	4.1% (51.3%)	0.0% (100.0%)	0.0% (100.0%)
	Post-2	9.4% (-11.1%)	9.2% (-9.5%)	0.0% (100.0%)	0.0% (100.0%)
95%	Pre-	8.4%	8.2%	0.4%	0.4%
	Post-1	3.9% (54.1%)	3.4% (58.9%)	0.0% (100.0%)	0.0% (100.0%)
	Post-2	9.9% (-17.9%)	9.5% (-15.9%)	0.0% (100.0%)	0.0% (100.0%)
Mean (10 Subjects)	Pre-	8.5%	7.9%	0.8%	0.8%
	Post-1	8.5% (-0.6%)	7.6% (3.7%)	0.0% (100.0%)	0.0% (100.0%)
	Post-2	8.3% (2.6%)	6.0% (24.3%)	0.0% (100.0%)	0.0% (100.0%)

Note:

Post-1 – Lifting assistance; Post-2 – Two-person team work; COMP – Disc compression force.

Table 40

Prevalences (Percent Reductions) of shoulder joint resultant forces above body weights and simultaneous shoulder muscle fatigue and shoulder joint resultant forces above body weights under different situations

Situation	FGH>BW	FGH>BW & Muscle Fatigue	FSC>BW	FSC>BW & Muscle Fatigue
Pre-	16.8%	5.1%	18.4%	5.2%
Post-1	15.9% (5.5%)	3.7% (28.4%)	17.9% (3.0%)	3.7% (28.5%)
Post-2	19.9% (-18.4%)	4.2% (18.8%)	23.2% (-25.8%)	4.2% (19.6%)

Note:

Post-1 – Lifting assistance; Post-2 – Two-person team work; FGH – Joint resultant force at glenohumeral; FSC – Joint resultant force at sternoclavicular; BW – Body weight.

Table 41

Number and average duration (Percent Reduction) of trunk muscle consecutive fatigue periods for different anthropometries under different situations

Parameter	Anthropometry	Pre-	Post-3	Post-1	Post-1&3	Post-2	Post-2&3
Number	Low (5%)	47	45 (4.3%)	27 (42.6%)	24 (48.9%)	78 (-66.0%)	70 (-48.9%)
	Average (50%)	29	22 (24.1%)	0 (100%)	0 (100%)	0 (100%)	0 (100%)
	High (95%)	9	7 (22.2%)	0 (100%)	0 (100%)	0 (100%)	0 (100%)
	Mean (10 subjects)	15	14 (6.7%)	6 (60.0%)	5 (66.7%)	9 (40.0%)	9 (40.0%)
Average Duration (Min)	Low (5%)	3.6	3.7 (-2.8%)	3.4 (5.6%)	3.4 (5.6%)	3.5 (2.8%)	3.5 (2.8%)
	Average (50%)	3.4	3.4 (0.0%)	0 (100%)	0 (100%)	0 (100%)	0 (100%)
	High (95%)	3.3	2.9 (12.1%)	0 (100%)	0 (100%)	0 (100%)	0 (100%)
	Mean (10 subjects)	3.7	3.6 (3.3%)	2.8 (25.5%)	2.7 (26.0%)	4.6 (-25.2%)	4.2 (-14.1%)

Note: Post-1 – Lifting assistance; Post-2 – Two-person team work; Post-3 – Work-rest schedule.

Table 42

Comparison and ranking of different intervention strategies (1 – most; 5 – least)

Situation	Low Back											
	Mean							Maximum			Rank_Low Back	
	MMCI	MF	JRF	Overload	Fatigue Rates	Consecutive Fatigue Periods	Fatigue & Overload	MMCI	MF	JRF	Sum of Score	Average
Post-1	1	1	1	3	2	2	3	3	3	3	22	0.44
Post-2	3	3	1	1	4	5	1	1	1	1	21	0.42
Post-3	5	5	5	5	5	3	5	5	5	5	48	0.96
Post-1&3	1	1	1	3	1	1	3	3	3	3	20	0.40
Post-2&3	3	3	1	1	3	5	1	1	1	1	20	0.40

Situation	Shoulder										Rank_Overall	
	Mean						Maximum		Rank_Shoulder			
	MF	JRF	Overload	Fatigue Rates	Consecutive Fatigue Periods	Fatigue & Overload	MF	JRF	Sum of Score	Average	Total	Average
Post-1	1	1	1	2	1	1	3	3	13	0.33	**35**	**0.39**
Post-2	1	1	5	4	1	3	1	1	17	0.43	**38**	**0.42**
Post-3	5	5	3	5	1	5	5	5	34	0.85	**82**	**0.91**
Post-1&3	1	1	1	1	1	1	3	3	12	0.30	**32**	**0.36**
Post-2&3	1	1	5	3	1	3	1	1	16	0.40	**36**	**0.40**

Note: MMCI – Maximum Muscle Contraction Intensity; MF – Muscle Forces; JRF – Joint Resultant Forces;
Post-1 – Lifting assistance; Post-2 – Two-person team work; Post-3 – Work-rest schedule.

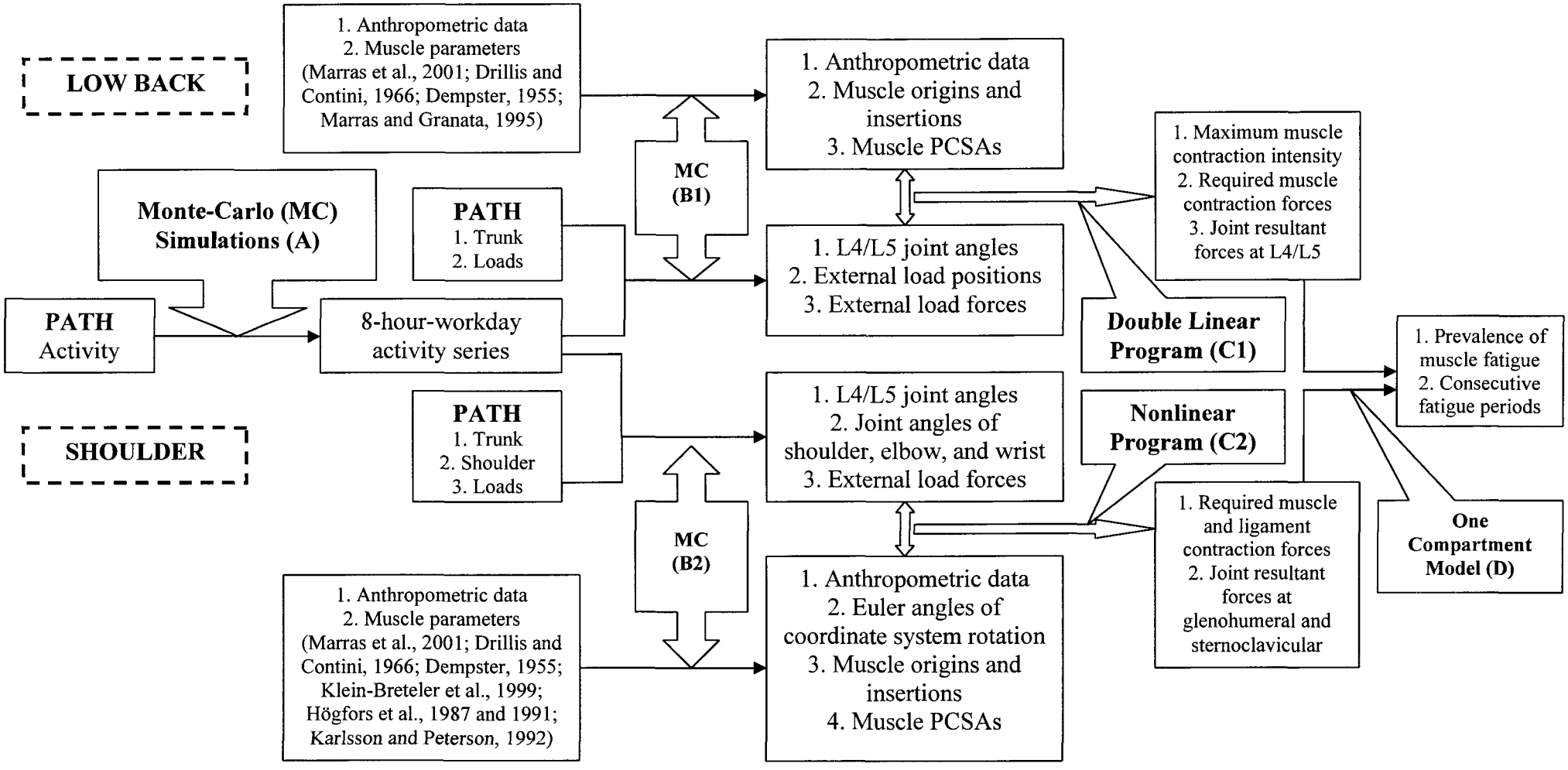

Figure 39 Schematic diagram for determination of dependent variables in the study

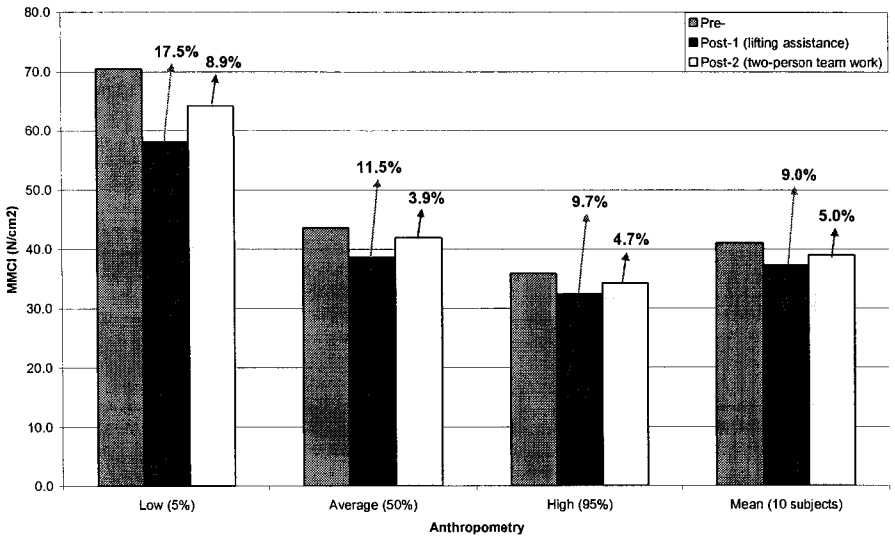

Figure 40 Comparisons of MMCI (Maximum Muscle Contraction Intensity) means (Percent Reductions) for different anthropometries during an 8-hour workday between post- and pre- interventions

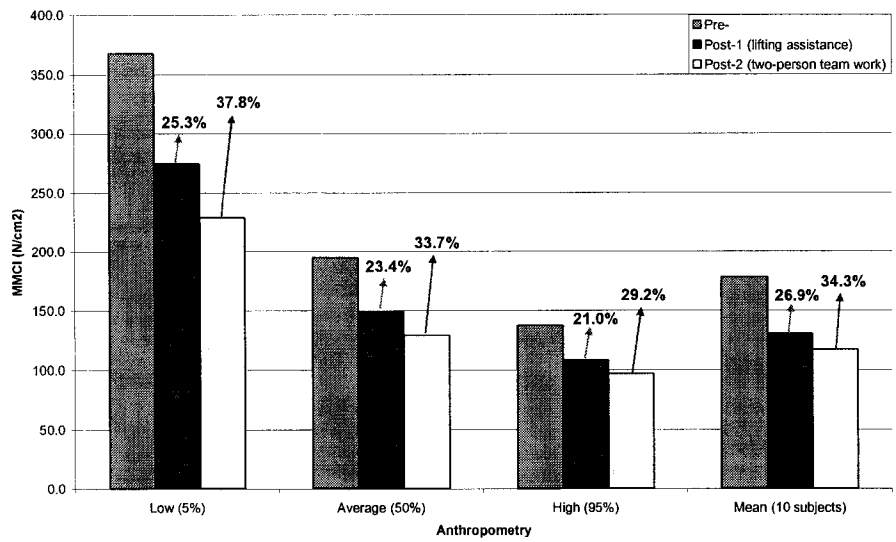

Figure 41 Comparisons of MMCI (Maximum Muscle Contraction Intensity) maximums (Percent Reductions) for different anthropometries during an 8-hour workday between post- and pre- interventions

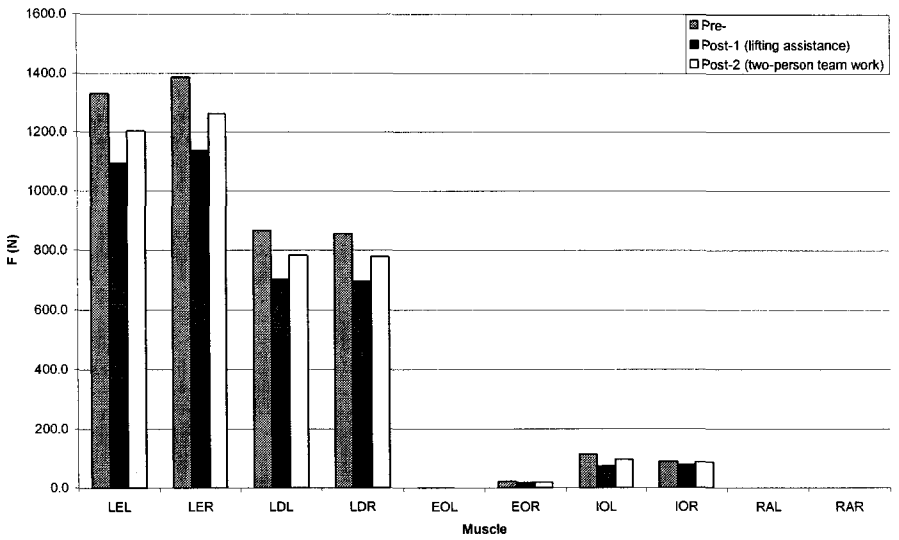

Figure 42 Comparisons of trunk muscle force means for 5 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

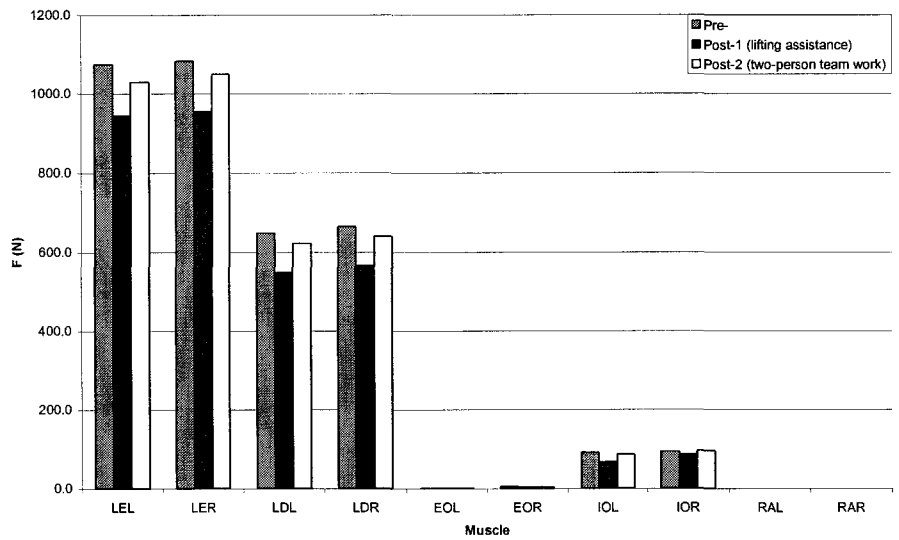

Figure 43 Comparisons of trunk muscle force means for 50 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

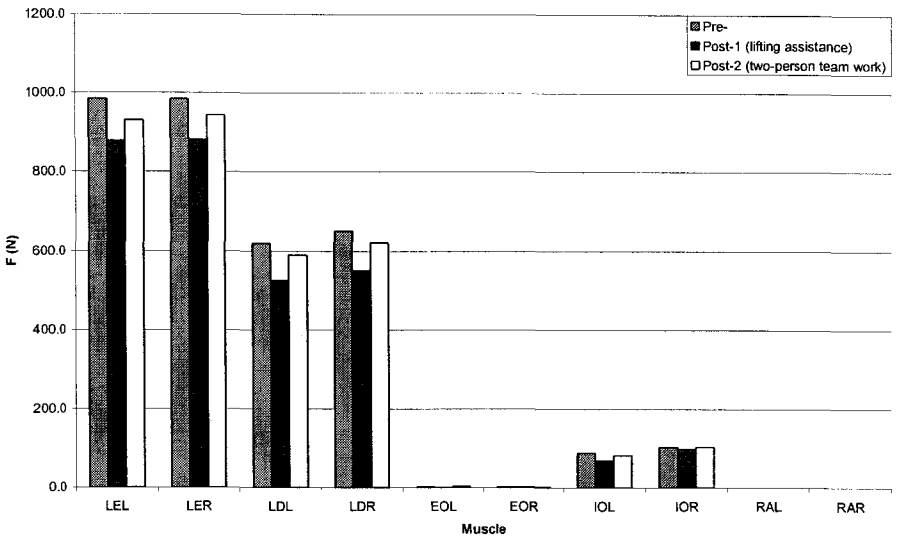

Figure 44 Comparisons of trunk muscle force means for 95 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

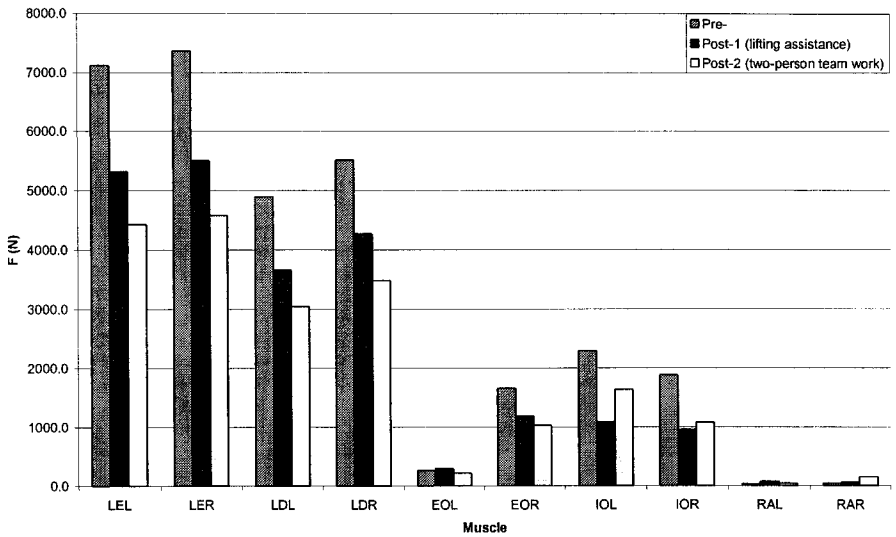

Figure 45 Comparisons of trunk muscle force maximums for 5 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

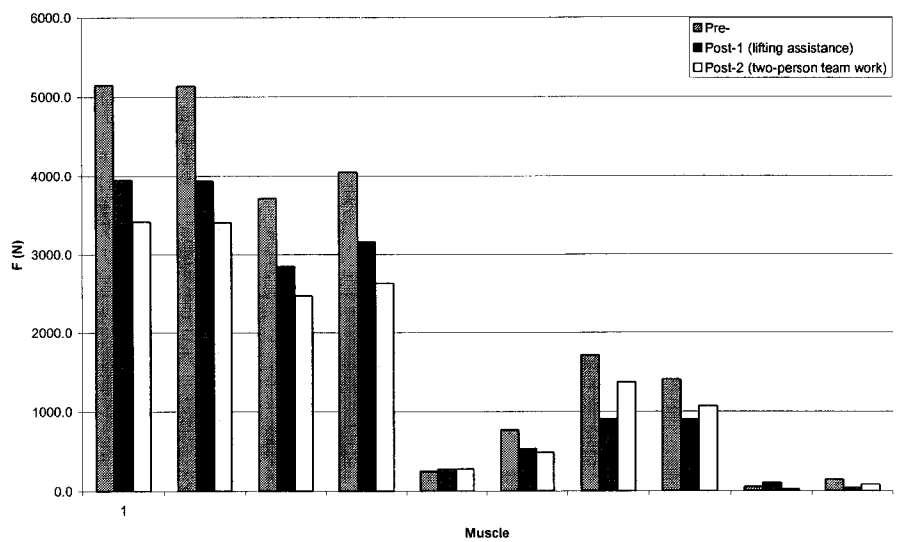

Figure 46 Comparisons of trunk muscle force maximums for 50 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

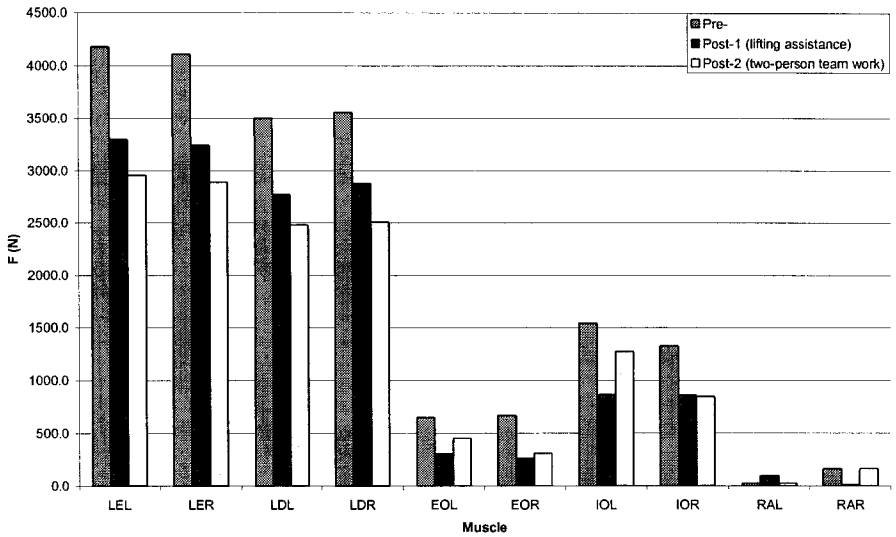

Figure 47 Comparisons of trunk muscle force maximums for 95 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

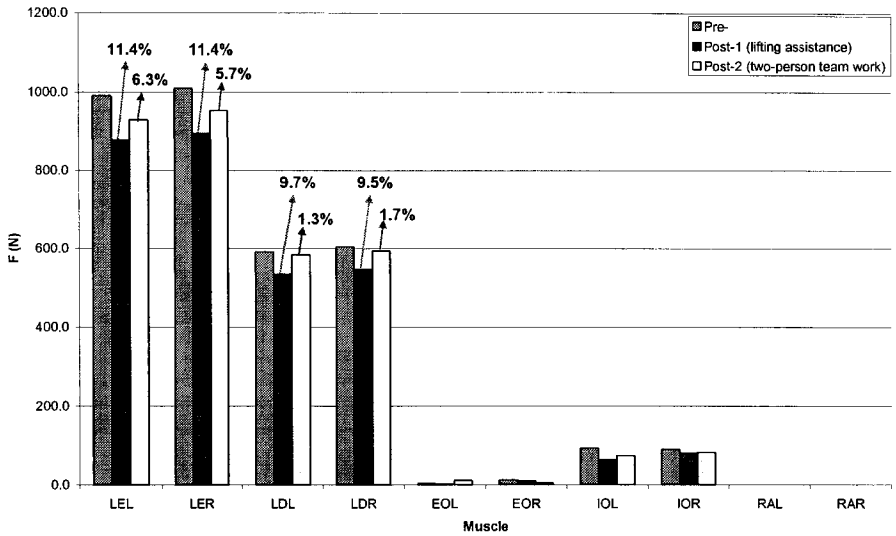

Figure 48 Comparisons of the averages of trunk muscle force means (Percent Reductions) for 10 simulated subjects during an 8-hour workday between post- and pre- interventions

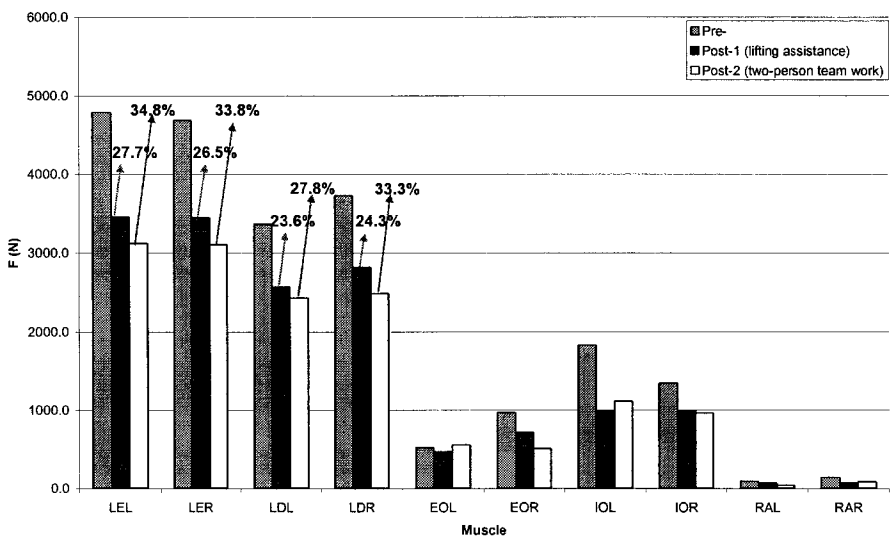

Figure 49 Comparisons of the averages of trunk muscle force maximums (Percent Reductions) for 10 simulated subjects during an 8-hour workday between post- and pre- interventions

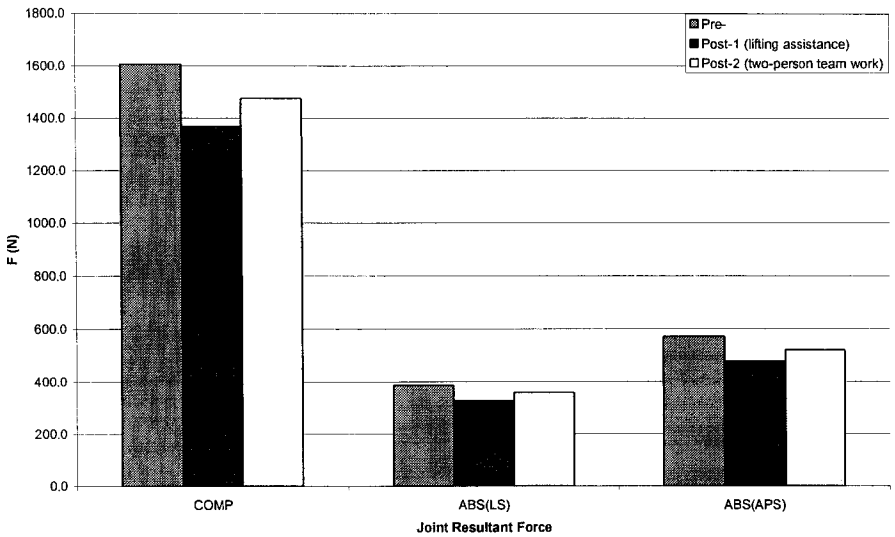

Figure 50 Comparisons of L4/L5 joint resultant force means for 5 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

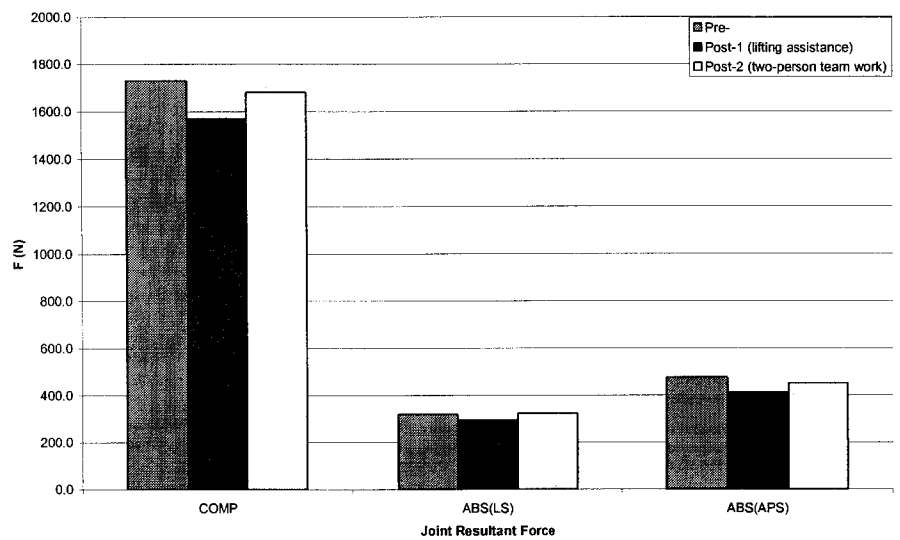

Figure 51 Comparisons of L4/L5 joint resultant force means for 50 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

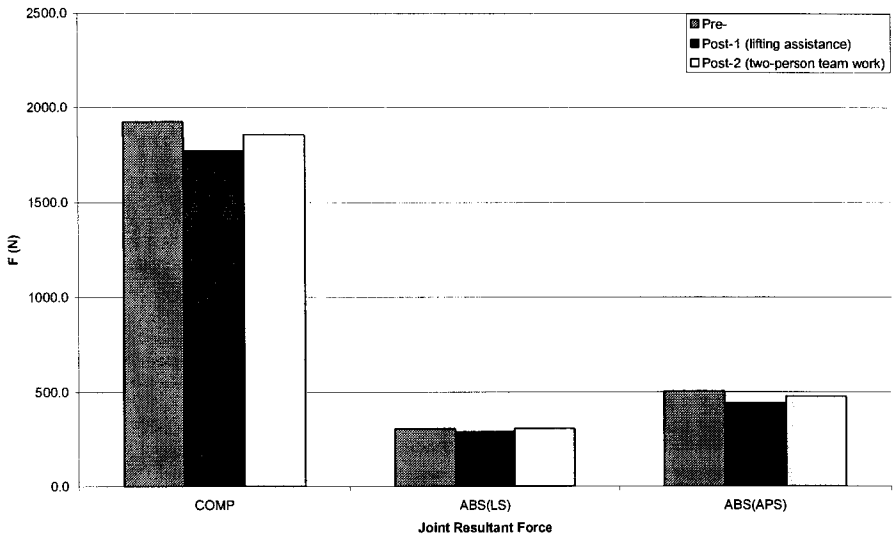

Figure 52 Comparisons of L4/L5 joint resultant force means for 95 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

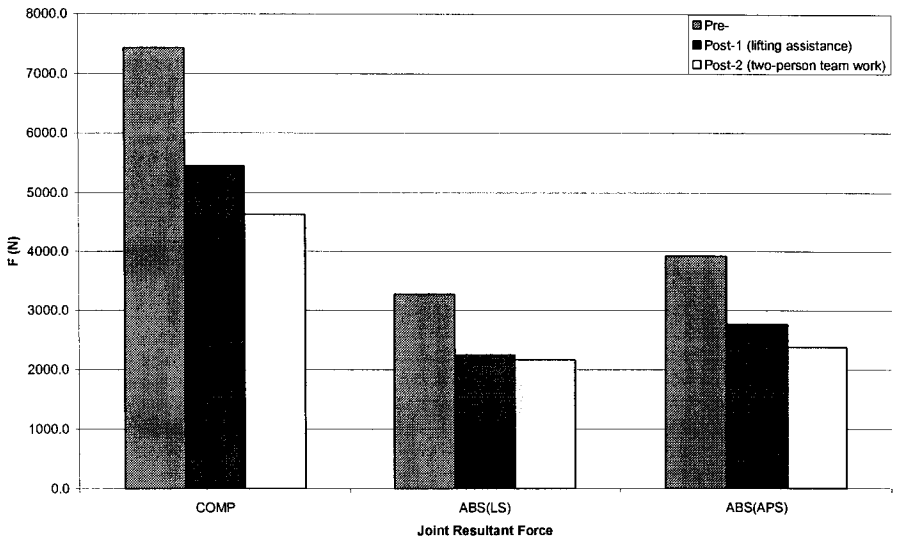

Figure 53 Comparisons of L4/L5 joint resultant force maximums for 5 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

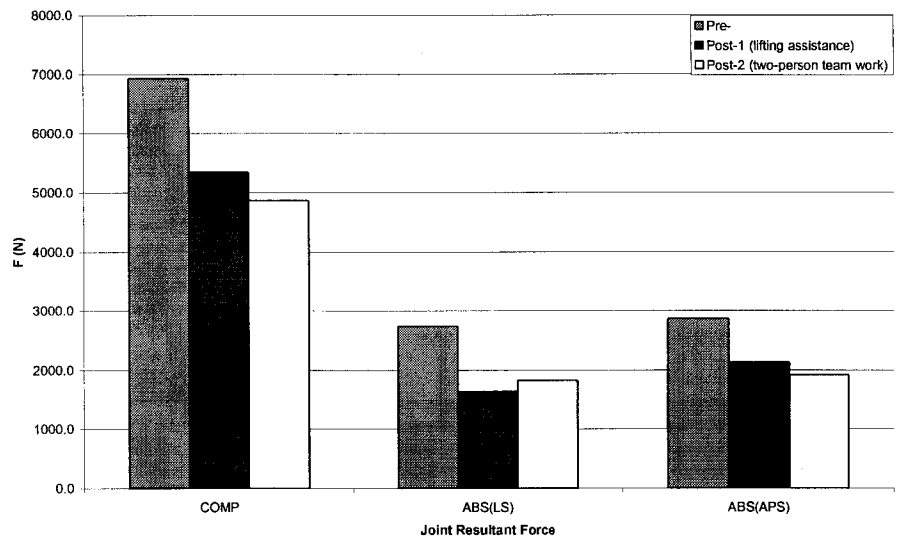

Figure 54 Comparisons of L4/L5 joint resultant force maximums for 50 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

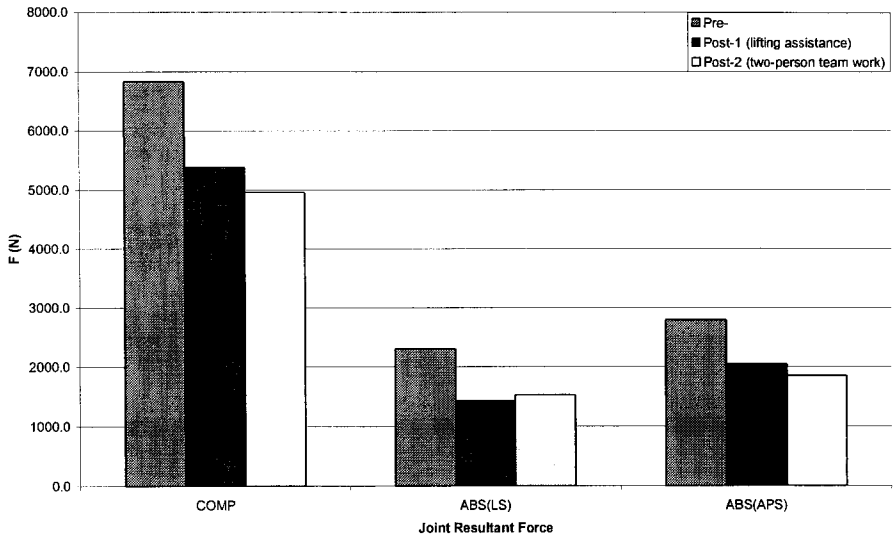

Figure 55 Comparisons of L4/L5 joint resultant force maximums for 95 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

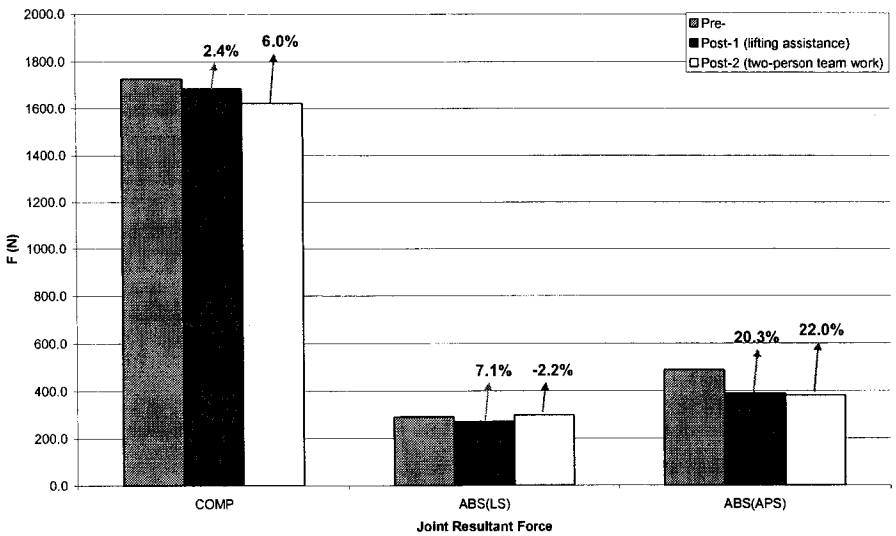

Figure 56 Comparisons of the averages of L4/L5 joint resultant force means (Percent Reductions) for 10 simulated subjects during an 8-hour workday between post- and pre- interventions

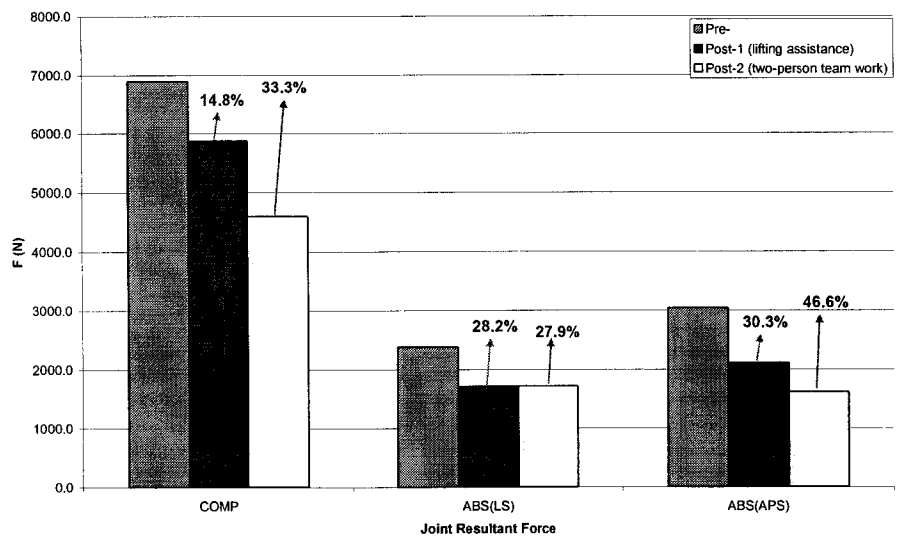

Figure 57 Comparisons of the averages of L4/L5 joint resultant force maximums (Percent Reductions) for 10 simulated subjects during an 8-hour workday between post- and pre- interventions

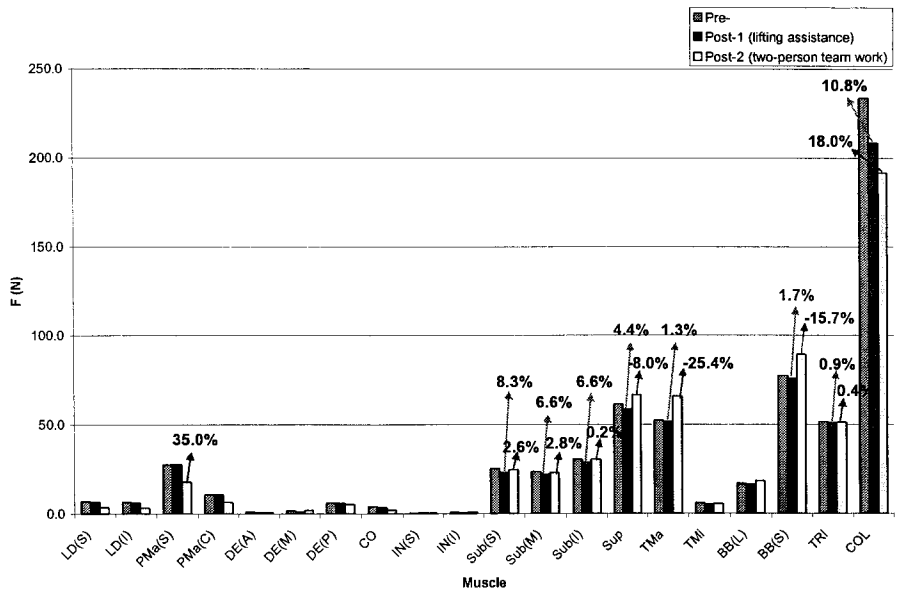

Figure 58 Comparisons of the averages of shoulder muscle force means (Percent Reductions) for 10 simulated subjects during an 8-hour workday between post- and pre- interventions

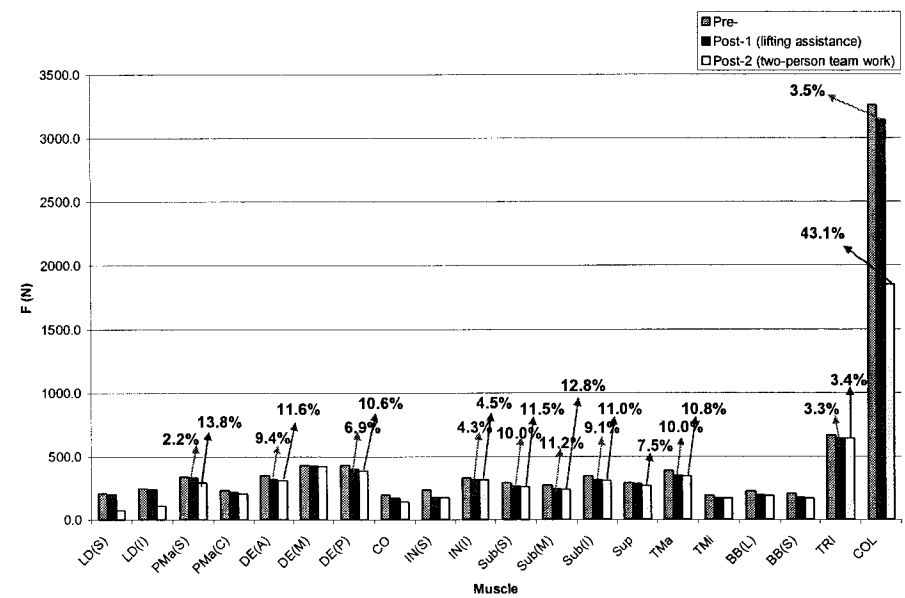

Figure 59 Comparisons of the averages of shoulder muscle force maximums (Percent Reductions) for 10 simulated subjects during an 8-hour workday between post- and pre- interventions

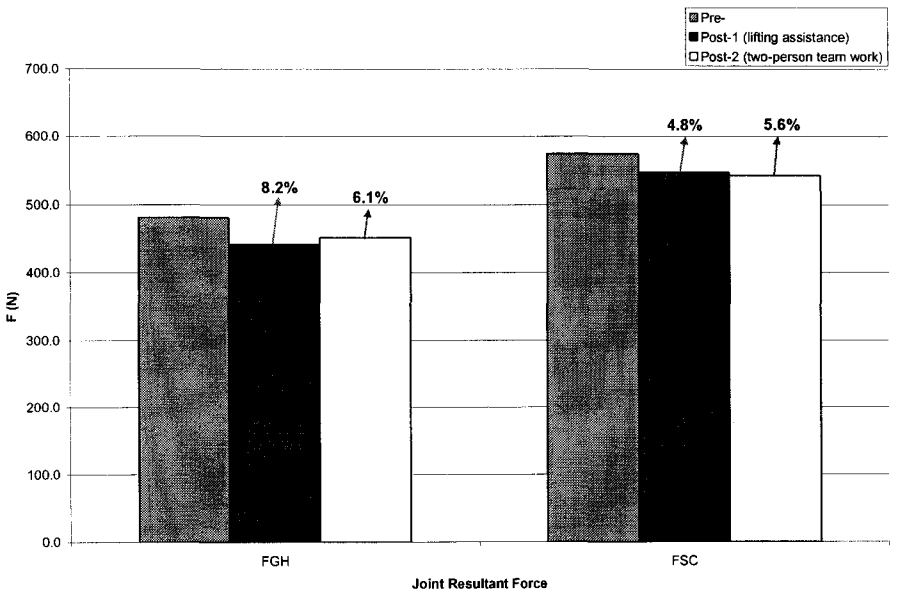

Figure 60 Comparisons of the averages of GH (glenohumeral) and SC (sternoclavicular) joint resultant force means (Percent Reductions) for 10 simulated subjects during an 8-hr workday between post- and pre- interventions

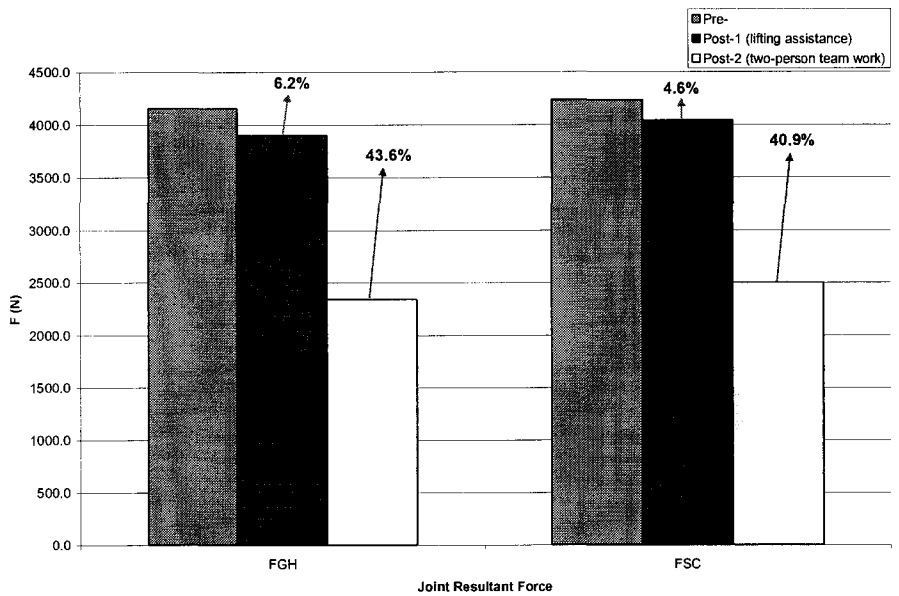

Figure 61 Comparisons of the averages of GH (glenohumeral) and SC (sternoclavicular) joint resultant force maximums (Percent Reductions) for 10 simulated subjects during an 8-hr workday between post- and pre- interventions

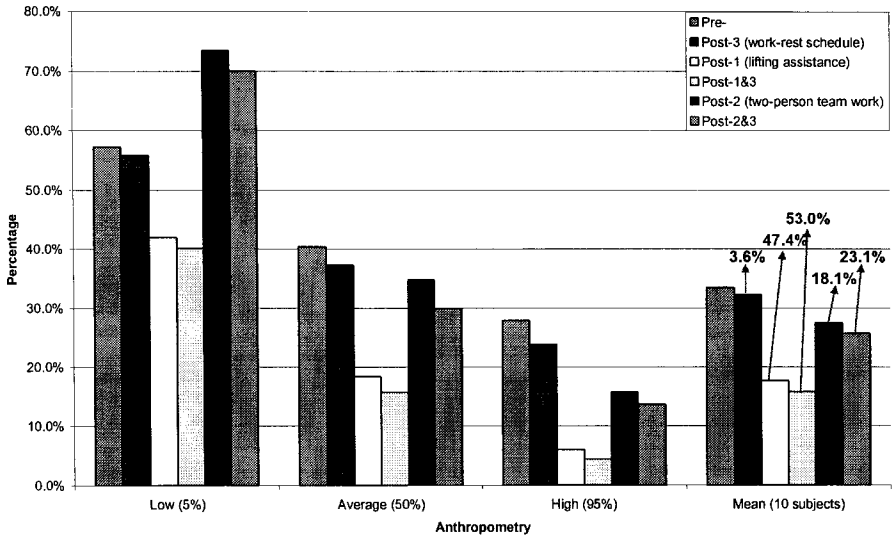

Figure 62 Comparisons of prevalences (Percent Reductions) of trunk muscle fatigue for different anthropometries during an 8-hour workday between post- and pre- interventions

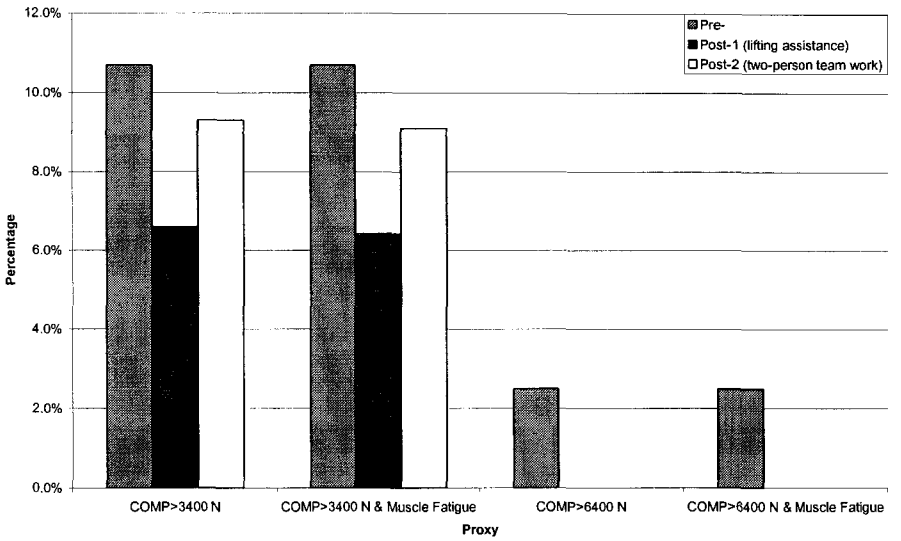

Figure 63 Comparisons of prevalences of trunk muscle fatigue and disc overload for 5 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

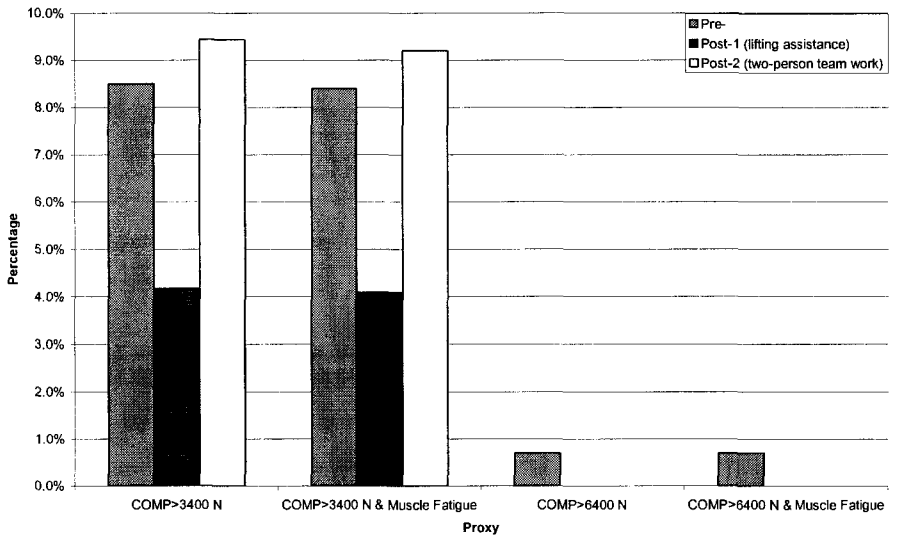

Figure 64 Comparisons of prevalences of trunk muscle fatigue and disc overload for 50 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

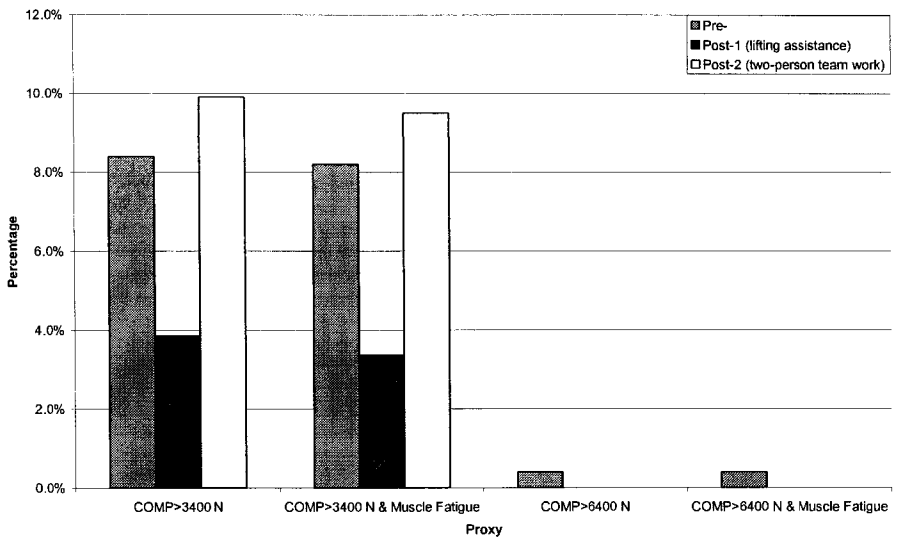

Figure 65 Comparisons of prevalences of trunk muscle fatigue and disc overload for 95 percentile of the hypothesized population during an 8-hour workday between post- and pre- interventions

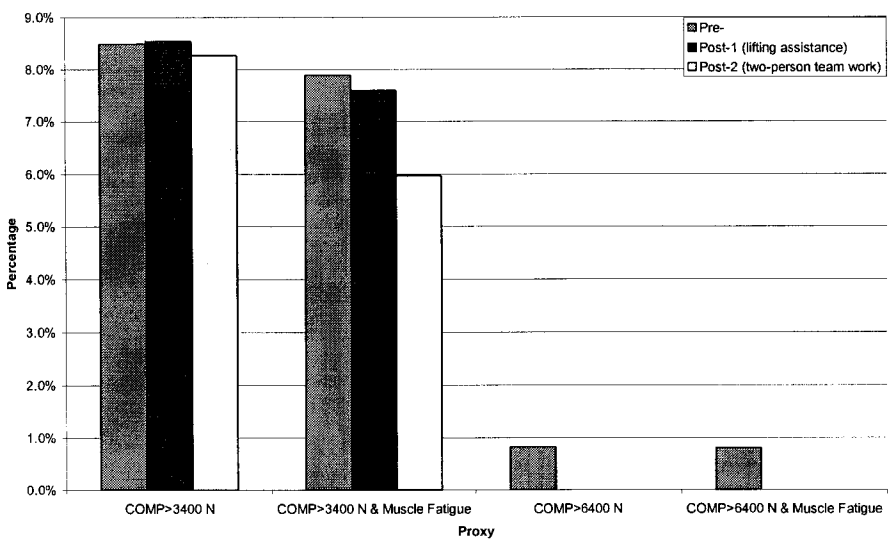

Figure 66 Comparisons of prevalences of trunk muscle fatigue and disc overload for 10 simulated subjects on average during an 8-hour workday between post- and pre- interventions

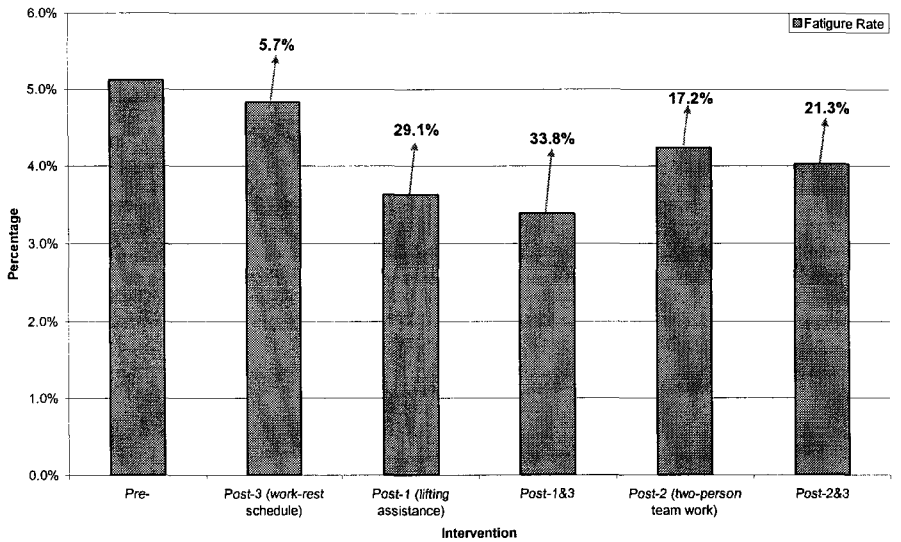

Figure 67 Comparisons of prevalences (Percent Reductions) of shoulder muscle fatigue for 10 simulated subjects on average during an 8-hour workday between post- and pre- interventions

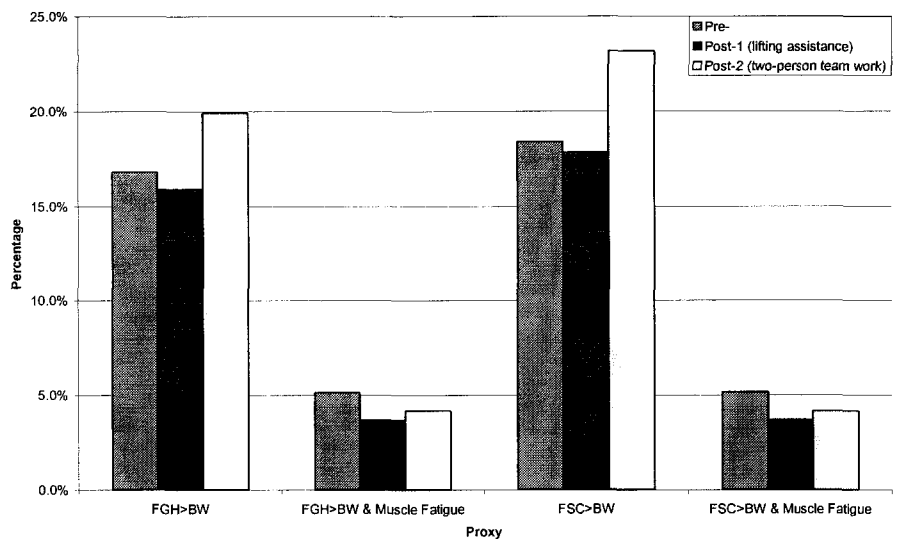

Figure 68 Comparisons of prevalences of shoulder muscle fatigue and joint overload for 10 simulated subjects on average during an 8-hour workday between post- and pre- interventions

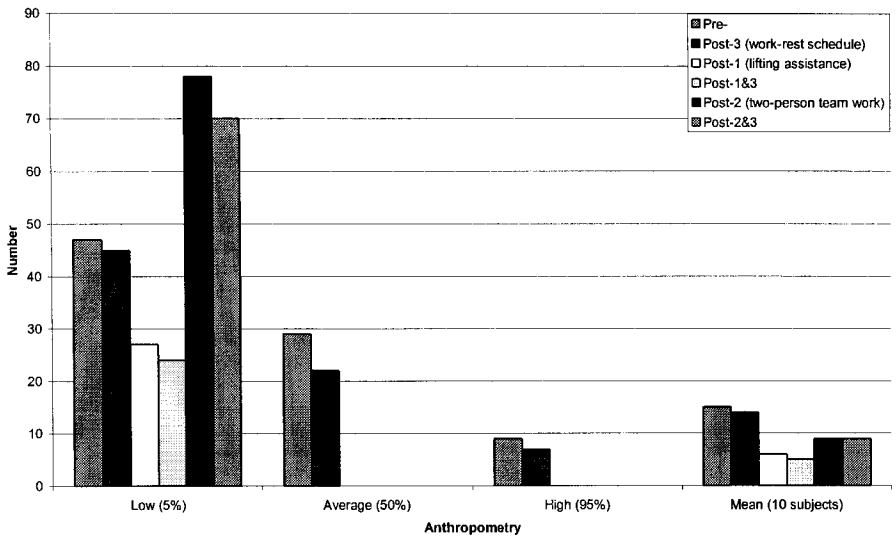

Figure 69 Comparisons of number of trunk muscle consecutive fatigue periods for different anthropometries during an 8-hour workday between post- and pre- interventions

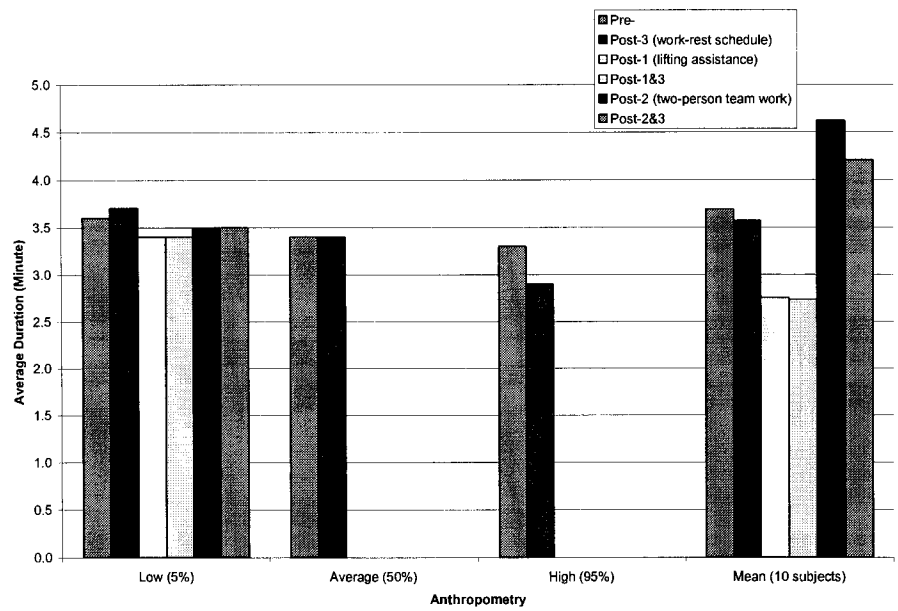

Figure 70 Comparisons of average duration of trunk muscle consecutive fatigue periods for different anthropometries during an 8-hour workday between post- and pre- interventions

V. Conclusion

This thesis provided a biomechanical analysis of the physical loads on the low back and shoulder during drywall installation longitudinally, integrating a variety of ergonomic exposure assessment techniques, PATH, Monte-Carlo simulation, biomechanical modeling of the low back and shoulder, and a one-compartment challenge-recovery model. It started with generating "typical" 8-hour-workday activity series by computer-aided simulation. Quantification of the ergonomic exposures then involved estimating the maximum muscle contraction intensity, required muscle contraction forces, and joint resultant forces of the low back and shoulder. Trunk and shoulder muscle fatigue were examined by computing the prevalences of muscle fatigue and consecutive fatigue periods during the simulated workday. In the end, ergonomic intervention strategies were proposed and investigated in terms of their efficacy.

The research suggested a cost-effective protocol to conducting both exposure assessment and intervention evaluation in the construction industry and other non-routine work sectors, where direct measurement technology is either infeasible or labor- and time- intensive to apply. It also emphasized the multifactorial determination of output values, which is driven by the variability of ergonomic exposures.

The validity of the biomechanical analysis was not directly evaluated due to lack of comparative studies, rather the integrated methods were aimed at the relative comparisons between the current ergonomic exposures and those under certain interventions. Of course, the overall validity of the thesis was threatened by several factors including the limited number of PATH observations, potential error during

random number generation, and various assumptions made during the construction of biomechanical modeling.

A. Biomechanical analysis of ergonomic exposure assessment

It was demonstrated that biomechanical models of the low back and shoulder developed by previous researchers could be used to estimate the physical loads for drywall installers, when the information about the model inputs was acquired by computer simulation based on the work-sampling based approach. The variability of exposure parameters including activities, postures, and weights handled was generated assuming certain distributions and thus impelled an ideal simulation of the reality.

More detailed and accurate data on both joint angles in three dimensions and muscle origin and insertion positions, however, are warranted for precise estimation of the muscle forces and joint resultant forces, especially at the shoulder. This would enrich the validity of biomechanical modeling over other exposure assessment approaches.

B. Multifactorial determination of biomechanical model outputs

Just as there are multiple risk factors for musculoskeletal injuries/disorders, a number of independent variables for the biomechanical models of the low back and shoulder, such as the subjects' anthropometric information, body part postures, loads in the hands, and the musculature-associated parameters (PCSA, origin, and insertion), etc., were necessary in terms of predicting the model output variables of muscle forces and joint resultant forces. Such a relationship was also surmised during shoulder analysis, where the results tended to be less varied compared to the low back analysis for 10

simulated subjects since only average information about the muscle parameters was available.

It is quite interesting and also challenging to examine the influence that different distribution types for the independent variables have on the results. Additionally, special efforts are encouraged to determine which variable is the most dominant one to affect the physical loads for the sake of effective intervention in the future.

C. Examination of muscle fatigue

By introducing the concept of muscle force production capacity, trunk and shoulder muscle fatigue during the simulated 8-hour drywall installation work was examined using a one-compartment challenge-recovery model. The prevalences of muscle fatigue, the number and average duration of consecutive fatigue periods were computed to illustrate the cumulative effects and the concentrated facts of muscle fatigue, respectively. The linkage between muscle fatigue and overload at the low back and shoulder joints was also investigated by pooling out the frequencies of simultaneous muscle fatigue and joint overload.

It was found in this study that the trunk muscles were more vulnerable to fatigue than the shoulder muscles. Given that the low back area is usually more frequently injured for drywall installers, such results may have provided more evidence to linking muscle fatigue and musculoskeletal injuries/disorders.

D. Strategy of intervention evaluation

The study evaluated the efficacy of ergonomic interventions for drywall installation using the integrated methods. Specific intervention strategies were proposed based on principles to insure their effectiveness. Subjects' anthropometric parameters and the 8-hour-workday activity series were randomized and matched with those values under pre- intervention. New input values under post- interventions were generated utilizing the similar methodologies as for ergonomic exposure assessment. Dependent variables for the biomechanical model and the one-compartment model, including the maximum muscle contraction intensity, muscle contraction forces, joint resultant forces, and muscle fatigue parameters, were regarded as intermediate outcomes to be evaluated and compared for the demonstration of intervention efficacy.

While the intervention of lifting assistance showed the advantages in reducing the average forces and muscle fatigue, the intervention of two-person team work was especially useful to lower the maximum force requirement. The intervention of optimal work-rest schedule also helped to attenuate muscle fatigue and when combined with the intervention of lifting assistance, became the relatively most efficacious one to reduce the physical ergonomic hazards during drywall installation.

E. Areas of future research

The research initiated insights into a significant evaluation of both the physical ergonomic exposures and occupational injury prevention interventions during drywall installation. It will benefit more from future studies in the following aspects:

1. An easy and cost-effective way for categorizing and collecting more detailed PATH posture data, especially for the shoulder movement.
2. Sensitivity analysis on simulations of independent variables assuming different distributions.
3. Determination of the most important contributor to intermediate outcomes during both exposure assessment and intervention evaluation.
4. Investigation of muscle fatigue variability in terms of which muscle fatigues and how it does.
5. Examination of intervention effectiveness in the long term integrating the considerations of different proxies.

Appendix A

PATH template for drywall installation

PATH Template for Drywall Installation (60 seconds)																					
Worker Number	1	2	3	4	5	6	7	8	9	10	**General Activities**	1	2	3	4	5	6	7	8	9	10
Worker											Climb/Descend										
Not Obs/Not Appl											Communicate										
Hand Posture	1	2	3	4	5	6	7	8	9	10	Cut										
Hand 1: Power Grasp											Fold										
Hand 1 : Pinch											In Between										
Hand 1: Other											Mark/Draw										
Hand 1: Empty											Measure										
Hand 1: Not Obs/Not Appl											Move Scaffold/Ladder										
Hand 2: Power Grasp											Whack										
Hand 2 : Pinch											Not Obs/Not Appl										
Hand 2: Other											**Tools**	1	2	3	4	5	6	7	8	9	10
Hand 2: Empty											Cutter										
Hand 2: Not Obs/Not Appl											Hammer										
Arm Posture	1	2	3	4	5	6	7	8	9	10	Level										
Both arms <60											Pencil										
1 arm ≥60											Rotary Zip										
2 arms ≥60											Screw Gun										
Not Obs/Not Appl											Sealant										
Trunk Posture	1	2	3	4	5	6	7	8	9	10	T-Square										
Neutral											Tape Measurer										
Mild Flexion											Not Obs/Not Appl										
Severe Flexion											**Equipment**	1	2	3	4	5	6	7	8	9	10
Bend/Twist											Cart										
Bend/Twist & Flexion											Ladder										
Not Obs/Not Appl											Scaffold										
Leg Posture	1	2	3	4	5	6	7	8	9	10	Not Obs/Not Appl										
Neutral											**Materials**	1	2	3	4	5	6	7	8	9	10
Kneel											1-in Sheetrock										
Knees Bent											5/8-in Sheetrock										
Legs Not Supported											Metal Studs										
Walk/Move											Nails/Screws										
Squat											Wires										
Not Obs/Not Appl											Not Obs/Not Appl										
Tasks	1	2	3	4	5	6	7	8	9	10	**MMH Activities**	1	2	3	4	5	6	7	8	9	10
Installing Sheetrocks											Carry										
Framing											Hold/Place										
Housekeeping											Lift										
Insulation											Lower/Throw										
Not Obs/Not Appl											Push/Pull/Drag										
Work Interface	1	2	3	4	5	6	7	8	9	10	Not Obs/Not Appl										
Bottom											**Weight in Hands (lbs)**	1	2	3	4	5	6	7	8	9	10
Top											Very Light										
Not Obs/Not Appl											Light										
Job Specific Activities	1	2	3	4	5	6	7	8	9	10	Medium										
Screw											Heavy										
Scrape Sheetrocks											Not Obs/Not Appl										
Load Gun																					
Not Obs/Not Appl																					

Appendix B

MATLAB double linear program to solve for trunk muscle forces and joint resultant forces at L4/L5

```
% Create an 8-hour-workday activity series;
for mm = 1:10
 for m = 1:100
   a0 = simdiscr(1,[0.12,0.10,0.16,0.10,0.05,0.04,0.06,0.05,0.08,0.06,0.10,0.08]);
    if a0 == 1
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0];
    elseif a0 == 2
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7];
    elseif a0 == 3
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0];
    elseif a0 == 4
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7];
    elseif a0 == 5
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 6
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    elseif a0 == 7
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 8
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    elseif a0 == 9
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 10
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    elseif a0 == 11
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 12
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    end
  end
end

% Generate reasonable values of joint angles, external load position and force vectors, and torque generated when
driving screw;
for nn = 1:10
 for n = 1:length(AS)
   ii(n,1,nn) = lognrnd(2.9,0.8); jj(n,1,nn) = randn*10.9+2.8; kk(n,1,nn) = randn*12.9-3.2;
   ez(n,1,nn) = randn*0.37+0.11;
   wy(n,1,nn) = 0; wz(n,1,nn) = -lognrnd(3.0,1.2);
   mtx(n,1,nn) = 0; mtz(n,1,nn) = 0;

   % trunk posture code 1 (neutral)
   ii1 = lognrnd(2.9,0.8);
   while ii1>20
     ii1 = lognrnd(2.9,0.8);
   end
   jj1 = randn*10.9+2.8;
```

```
while abs(jj1)>20
 jj1 = randn*10.9+2.8;
end
kk1 = randn*12.9-3.2;
while abs(kk1)>20
 kk1 = randn*12.9-3.2;
end
ez1 = randn*0.37+0.11;
while (ez1<0)|(ez1>0.8)
 ez1 = randn*0.37+0.11;
end

% trunk posture code 2 (mild flexion)
ii2 = lognrnd(2.9,0.8);
while (ii2<=20)|(ii2>45)
 ii2 = lognrnd(2.9,0.8);
end
jj2 = jj1;
kk2 = kk1;
ez2 = randn*0.37+0.11;
while (ez2<-0.4)|(ez2>0.2)
 ez2 = randn*0.37+0.11;
end

% trunk posture code 3 (severe flexion)
ii3 = lognrnd(2.9,0.8);
while (ii3<=45)|(ii3>80)
 ii3 = lognrnd(2.9,0.8);
end
jj3 = jj1;
kk3 = kk1;
ez3 = randn*0.37+0.11;
while (ez3<-0.9)|(ez3>-0.1)
 ez3 = randn*0.37+0.11;
end

% trunk posture code 4 (bend & twist)
ii4 = ii1;
jj41 = randn*10.9+2.8;
jj42 = randn*10.9+2.8;
kk41 = randn*12.9-3.2;
kk42 = randn*12.9-3.2;
PL = 0.8;
PPL = (rand<=PL);
if PPL == 0
 while (jj41<-35)|(jj41>-20)
    jj41 = randn*10.9+2.8;
 end
 while (kk41<20)|(kk41>45)
    kk41 = randn*12.9-3.2;
 end
elseif PPL == 1
 while (jj42<20)|(jj42>35)
    jj42 = randn*10.9+2.8;
 end
 while (kk42<-45)|(kk42>-20)
    kk42 = randn*12.9-3.2;
 end
end
```

```
ez4 = ez1;

% trunk posture code 5 (bend & twist & flexion)
ii5 = lognrnd(2.9,0.8);
while (ii5<=20)|(ii5>80)
  ii5 = lognrnd(2.9,0.8);
end
jj51 = jj41;
jj52 = jj42;
kk51 = kk41;
kk52 = kk42;
ez5 = randn*0.37+0.11;
while (ez5<-0.9)|(ez5>0.2)
  ez5 = randn*0.37+0.11;
end

switch AS(n,1,nn)
 case 0
  ii(n,1,nn) = 0; jj(n,1,nn) = 0; kk(n,1,nn) = 0;
  ex(n,1,nn) = 0; ey(n,1,nn) = 0; ez(n,1,nn) = 0;
  wx(n,1,nn) = 0; wy(n,1,nn) = 0; wz(n,1,nn) = 0;
  mtx(n,1,nn) = 0; mty(n,1,nn) = 0; mtz(n,1,nn) = 0;

 case 1 % activity 1 (cut/measure)
  ex(n,1,nn) = 0.3; wx(n,1,nn) = 10; wz(n,1,nn) = -2; mty(n,1,nn) = 0;
  c1 = simdiscr(1,[12,4,3,1,5]/25);
  if c1 == 1 % trunk posture code 1 (neutral)
    ey(n,1,nn) = 0.3; ez(n,1,nn) = ez1;
    ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

  elseif c1 == 2 % trunk posture code 2 (mild flexion)
    ey(n,1,nn) = 0.4; ez(n,1,nn) = ez2;
    ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

  elseif c1 == 3 % trunk posture code 3 (severe flexion)
    ey(n,1,nn) = 0.5; ez(n,1,nn) = ez3;
    ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;

  elseif c1 == 4 % trunk posture code 4 (bend & twist)
    ey(n,1,nn) = 0.3; ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
    elseif PPL == 1
      jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
    end

  elseif c1 == 5 % trunk posture code 5 (bend & twist & flexion)
    ey(n,1,nn) = 0.4; ez(n,1,nn) = ez5; ii(n,1,nn) = ii5;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
    elseif PPL == 1
      jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
    end
  end
```

```
case 2 % activity 2 (lift)
 ex(n,1,nn) = 0; ey(n,1,nn) = 0.6; wx(n,1,nn) = 0; wz(n,1,nn) = -408; mty(n,1,nn) = 0;
 c2 = simdiscr(1,[0,2,1,0,5]/8);
 if c2 == 1 % trunk posture code 1 (neutral)
  ez(n,1,nn) = ez1;
  ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

 elseif c2 == 2 % trunk posture code 2 (mild flexion)
  ez(n,1,nn) = ez2;
  ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

 elseif c2 == 3 % trunk posture code 3 (severe flexion)
  ez(n,1,nn) = ez3;
  ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;

 elseif c2 == 4 % trunk posture code 4 (bend & twist)
  ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
   jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
  elseif PPL == 1
   jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
  end

 elseif c2 == 5 % trunk posture code 5 (bend & twist & flexion)
  ez(n,1,nn) = ez3; ii(n,1,nn) = ii5;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
   jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
  elseif PPL == 1
   jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
  end
 end

case 3 % activity 3 (carry)
 ex(n,1,nn) =0; wx(n,1,nn) = 0; mty(n,1,nn) = 0;
 if (AS(n-1,1,nn) == 2)|(AS(n-2,1,nn) == 2)|(AS(n-3,1,nn) == 2)|(AS(n-4,1,nn) == 2)|(AS(n-5,1,nn) == 2)
   wz(n,1,nn) = -408;
 else
   wz1 = -lognrnd(3.0,1.2);
    while wz1<-23
     wz1 = -lognrnd(3.0,1.2);
    end
   wz2 = -lognrnd(3.0,1.2);
    while (wz2<-67)|(wz2>-23)
     wz2 = -lognrnd(3.0,1.2);
    end
   wz3 = -lognrnd(3.0,1.2);
    while (wz3<-223)|(wz3>-67)
     wz3 = -lognrnd(3.0,1.2);
    end
   wz4 = -lognrnd(3.0,1.2);
    while (wz4<-408)|(wz4>-223)
     wz4 = -lognrnd(3.0,1.2);
    end
   c33 = simdiscr(1,[1,5,4,6]/16);
   if c33 == 1
```

```
      wz(n,1,nn) = wz1;
    elseif c33 == 2
      wz(n,1,nn) = wz2;
    elseif c33 == 3
      wz(n,1,nn) = wz3;
    elseif c33 == 4
      wz(n,1,nn) = wz4;
    end
  end

  c3 = simdiscr(1,[15,0,1,0,0]/16);
  if c3 == 1 % trunk posture code 1 (neutral)
    ey(n,1,nn) = 0.3; ez(n,1,nn) = ez1;
    ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

  elseif c3 == 2 % trunk posture code 2 (mild flexion)
    ey(n,1,nn) = 0.4; ez(n,1,nn) = ez2;
    ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

  elseif c3 == 3 % trunk posture code 3 (severe flexion)
    ey(n,1,nn) = 0.5; ez(n,1,nn) = ez3;
    ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;

  elseif c3 == 4 % trunk posture code 4 (bend & twist)
    ey(n,1,nn) = 0.3; ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
    elseif PPL == 1
      jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
    end

  elseif c3 == 5 % trunk posture code 5 (bend & twist & flexion)
    ey(n,1,nn) = 0.4; ez(n,1,nn) = ez5; ii(n,1,nn) = ii5;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
    elseif PPL == 1
      jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
    end
  end

case 4 % activity 4 (hold/place)
  ex(n,1,nn) = 0; wx(n,1,nn) = 0; wz(n,1,nn) = -408; mty(n,1,nn) = 0;
  c4 = simdiscr(1,[0.33,0.17,0.33,0.17,0]);
  if c4 == 1 % trunk posture code 1 (neutral)
    ey(n,1,nn) = 0.2; ez(n,1,nn) = ez1;
    ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

  elseif c4 == 2 % trunk posture code 2 (mild flexion)
    ey(n,1,nn) = 0.3; ez(n,1,nn) = ez2;
    ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

  elseif c4 == 3 % trunk posture code 3 (severe flexion)
    ey(n,1,nn) = 0.4; ez(n,1,nn) = ez3;
    ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;
```

```
 elseif c4 == 4 % trunk posture code 4 (bend & twist)
  ey(n,1,nn) = 0.2; ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
  elseif PPL == 1
    jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
  end

 elseif c4 == 5 % trunk posture code 5 (bend & twist & flexion)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez5; ii(n,1,nn) = ii5;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
  elseif PPL == 1
    jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
  end
 end

case 5 % activity 5 (screw)
 ex(n,1,nn) = 0.2; wx(n,1,nn) = 0; wy(n,1,nn) = -10; wz(n,1,nn) = -12; mty(n,1,nn) = -7;
 c5 = simdiscr(1,[7,5,3,3,2]/20);
 if c5 == 1 % trunk posture code 1 (neutral)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez1;
  ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

 elseif c5 == 2 % trunk posture code 2 (mild flexion)
  ey(n,1,nn) = 0.4; ez(n,1,nn) = ez2;
  ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

 elseif c5 == 3 % trunk posture code 3 (severe flexion)
  ey(n,1,nn) = 0.5; ez(n,1,nn) = ez3;
  ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;

 elseif c5 == 4 % trunk posture code 4 (bend & twist)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
  elseif PPL == 1
    jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
  end

 elseif c5 == 5 % trunk posture code 5 (bend & twist & flexion)
  ey(n,1,nn) = 0.4; ez(n,1,nn) = ez5; ii(n,1,nn) = ii5;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
  elseif PPL == 1
    jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
  end
 end

case 6 % activity 6 (in between)
 ex(n,1,nn) = 0; wx(n,1,nn) = 0; wz(n,1,nn) = -12; mty(n,1,nn) = 0;
```

```
c6 = simdiscr(1,[22,2,0,0,0]/24);
if c6 == 1 % trunk posture code 1 (neutral)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez1;
  ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

 elseif c6 == 2 % trunk posture code 2 (mild flexion)
  ey(n,1,nn) = 0.4; ez(n,1,nn) = ez2;
  ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

 elseif c6 == 3 % trunk posture code 3 (severe flexion)
  ey(n,1,nn) = 0.5; ez(n,1,nn) = ez3;
  ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;

 elseif c6 == 4 % trunk posture code 4 (bend & twist)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
  elseif PPL == 1
    jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
  end

 elseif c6 == 5 % trunk posture code 5 (bend & twist & flexion)
  ey(n,1,nn) = 0.4; ez(n,1,nn) = ez5; ii(n,1,nn) = ii5;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
  elseif PPL == 1
    jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
  end
 end

case 7 % activity 7 (other)
 ex(n,1,nn) = 0; wx(n,1,nn) = 0; mty(n,1,nn) = 0;
 while wz(n,1,nn)<-23
    wz(n,1,nn) = -lognrnd(3.0,1.2);
 end
 c7 = simdiscr(1,[13,2,8,0,4]/27);
 if c7 == 1 % trunk posture code 1 (neutral)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez1;
  ii(n,1,nn) = ii1; jj(n,1,nn) = jj1; kk(n,1,nn) = kk1;

 elseif c7 == 2 % trunk posture code 2 (mild flexion)
  ey(n,1,nn) = 0.4; ez(n,1,nn) = ez2;
  ii(n,1,nn) = ii2; jj(n,1,nn) = jj2; kk(n,1,nn) = kk2;

 elseif c7 == 3 % trunk posture code 3 (severe flexion)
  ey(n,1,nn) = 0.5; ez(n,1,nn) = ez3;
  ii(n,1,nn) = ii3; jj(n,1,nn) = jj3; kk(n,1,nn) = kk3;

 elseif c7 == 4 % trunk posture code 4 (bend & twist)
  ey(n,1,nn) = 0.3; ez(n,1,nn) = ez4; ii(n,1,nn) = ii4;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj(n,1,nn) = jj41; kk(n,1,nn) = kk41;
  elseif PPL == 1
```

```
      jj(n,1,nn) = jj42; kk(n,1,nn) = kk42;
    end

  elseif c7 == 5 % trunk posture code 5 (bend & twist & flexion)
    ey(n,1,nn) = 0.4; ez(n,1,nn) = ez5; ii(n,1,nn) = ii5;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj(n,1,nn) = jj51; kk(n,1,nn) = kk51;
    elseif PPL == 1
      jj(n,1,nn) = jj52; kk(n,1,nn) = kk52;
    end
  end
 end
 U(:,:,nn) = [AS(:,nn) ii(:,nn) jj(:,nn) kk(:,nn) ex(:,nn) ey(:,nn) ez(:,nn) wx(:,nn) wy(:,nn) wz(:,nn) mtx(:,nn) mty(:,nn)
mtz(:,nn)];
 AN(:,:,nn)= [ii(:,nn) jj(:,nn) kk(:,nn)];
 EL(:,:,nn) = [ex(:,nn) ey(:,nn) ez(:,nn)];
 FEL(:,:,nn) = [wx(:,nn) wy(:,nn) wz(:,nn)];
 MT(:,:,nn) = [mtx(:,nn) mty(:,nn) mtz(:,nn)];
end

% Obtain a series of internal moment vectors B;
for qq = 1:10
  H(qq) = randn*9.1+175.9;
  while (H(qq)<160)|(H(qq)>200)
    H(qq) = randn*9.1+175.9;
  end
  HEIGHT(qq) = H(qq)/100; UB(:,qq) = [0;0;HEIGHT(qq)*0.47*0.396];
  W(qq) = randn*13.3+79.8;
  while (W(qq)<65)|(W(qq)>110)
    W(qq) = randn*13.3+79.8;
  end
  WEIGHT(qq) = W(qq)*9.8; FUB(:,qq) = [0;0;-WEIGHT(qq)*35.8/75.2];
end

for rr = 1:10
  ub = UB(:,rr); fub = FUB(:,rr);
 for r = 1:length(AS)
  ii = AN(r,1,rr); jj = AN(r,2,rr); kk = AN(r,3,rr); el = EL(r,:,rr)'; fel = FEL(r,:,rr)'; t = MT(r,:,rr)';
  i = ii*pi/180; j = jj*pi/180; k = kk*pi/180;
  T = [      cos(j)*cos(k)                    cos(j)*sin(k)                -sin(j)
     -cos(i)*sin(k)+sin(i)*sin(j)*cos(k)   cos(i)*cos(k)+sin(i)*sin(j)*sin(k)   sin(i)*cos(j)
      sin(i)*sin(k)+cos(i)*sin(j)*cos(k)   -sin(i)*cos(k)+cos(i)*sin(j)*sin(k)   cos(i)*cos(j)];
  fub1 = T'*fub; el1 = T'*el; fel1 = T'*fel; t1 = T'*t;
  fe = fub1+fel1; me = cross(ub,fub1)+cross(el1,fel1)+t1;
  G(:,r,rr) = fe; B(:,r,rr) = -me;
 end
end

% Obtain muscle force vector F and muscle-force-generated moment vector M;
for ss = 1:10
  TW1(ss) = randn*0.022+0.303; TD1(ss) = randn*0.022+0.223;
  TW2(ss) = randn*0.020+0.324; TD2(ss) = randn*0.022+0.229; Z(ss) = -0.0275*HEIGHT(ss)+0.3*TD2(ss);
  coe1 = [TW1(ss)  0  0; 0 TD1(ss)  0; 0  0  1];
  coe2 = [TW2(ss)  0  0; 0 TD2(ss)  0; 0  0  Z(ss)];
  ca1 = [-0.2;-0.3;0];  cb1 = [-0.3;-0.3;1];   a1 = coe1*ca1; b1 = coe2*cb1; c1 = (a1-b1)/norm(a1-b1); m1 =
cross(a1,c1);
```

```
   ca2 = [0.2;-0.3;0];   cb2 = [0.3;-0.3;1];    a2 = coe1*ca2; b2 = coe2*cb2; c2 = (a2-b2)/norm(a2-b2); m2 =
cross(a2,c2);
   ca3 = [-0.25;-0.3;0]; cb3 = [-0.6;0.1;1];    a3 = coe1*ca3; b3 = coe2*cb3; c3 = (a3-b3)/norm(a3-b3); m3 =
cross(a3,c3);
   ca4 = [0.25;-0.3;0];  cb4 = [0.6;0.1;1];     a4 = coe1*ca4; b4 = coe2*cb4; c4 = (a4-b4)/norm(a4-b4); m4 =
cross(a4,c4);
   ca5 = [-0.1;0.55;0];  cb5 = [-0.45;-0.19;1]; a5 = coe1*ca5; b5 = coe2*cb5; c5 = (a5-b5)/norm(a5-b5); m5 =
cross(a5,c5);
   ca6 = [0.1;0.55;0];   cb6 = [0.45;-0.19;1];  a6 = coe1*ca6; b6 = coe2*cb6; c6 = (a6-b6)/norm(a6-b6); m6 =
cross(a6,c6);
   ca7 = [-0.45;0.3;0];  cb7 = [-0.45;0.2;1];   a7 = coe1*ca7; b7 = coe2*cb7; c7 = (a7-b7)/norm(a7-b7); m7 =
cross(a7,c7);
   ca8 = [0.45;0.3;0];   cb8 = [0.45;0.2;1];    a8 = coe1*ca8; b8 = coe2*cb8; c8 = (a8-b8)/norm(a8-b8); m8 =
cross(a8,c8);
   ca9 = [-0.1;0.55;0];  cb9 = [-0.1;0.55;1];   a9 = coe1*ca9; b9 = coe2*cb9; c9 = (a9-b9)/norm(a9-b9); m9 =
cross(a9,c9);
   ca10 = [0.1;0.55;0];  cb10 = [0.1;0.55;1];   a10 = coe1*ca10; b10 = coe2*cb10; c10 = (a10-b10)/norm(a10-b10);
m10 = cross(a10,c10);
   F(:,:,ss) = [c1 c2 c3 c4 c5 c6 c7 c8 c9 c10];
   M(:,:,ss) = [m1 m2 m3 m4 m5 m6 m7 m8 m9 m10];
end

%  Execute the Double Linear Optimization Program to obtain a series of maximum muscle force intensity, muscle
forces, and joint resultant forces.
f = [1,1,1,1,1,1,1,1,1,1];
vlb = [0;0;0;0;0;0;0;0;0;0];
for tt = 1:10
   p1 = 53.65-12.34*H(tt)/W(tt); p2 = 50.7-11.04*H(tt)/W(tt); p3 = -4.65+0.032*TW2(tt)*TD2(tt)*10000; p4 =
3.44+0.229*W(tt); p5 = 0.315+0.014*TW2(tt)*TD2(tt)*10000;
   p6 = 1.2+0.013*TW2(tt)*TD2(tt)*10000; p7 = -4.05+32.29*W(tt)/H(tt); p8 = 1.01+0.116*W(tt); p9 = 24.44-
6.87*H(tt)/W(tt); p10 = 23.84-6.6*H(tt)/W(tt);
   PCSA(:,tt) = [p1;p2;p3;p4;p5;p6;p7;p8;p9;p10];
   V(:,:,tt) = [H(tt) W(tt) TW1(tt) TD1(tt) TW2(tt) TD2(tt)];
end

for uu = 1:10
   A = M(:,:,uu); FF = F(:,:,uu); pcsa = PCSA(:,uu);
  for u = 1:length(AS)
   b = B(:,u,uu); fe = G(:,u,uu); UU = U(u,:,uu);
     l = 22;
     z = 'in';
   while z == 'in'
     l = l + 0.1;
     vub = l.*pcsa;
     [x,y,z] = lp(f,A,b,vlb,vub,[],3);
     z = [z(1,1),z(1,2)];
     lmax = x./pcsa;
     N = -FF*x-fe; X = [l;x;N]';
     ANSWER(u,:,uu) = [UU X];
   end
  end
end
```

Appendix C

MATLAB nonlinear optimization program to solve for shoulder muscle forces and resultant forces at GH (glenohumeral) and SC (sternoclavicular) joints

```
% Create an 8-hour-workday activity series;
for mm = 1:10
 for m = 1:100
   a0 = simdiscr(1,[0.12,0.10,0.16,0.10,0.05,0.04,0.06,0.05,0.08,0.06,0.10,0.08]);
    if a0 == 1
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0];
    elseif a0 == 2
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7];
    elseif a0 == 3
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0];
    elseif a0 == 4
     AS((55*(m-1)+1):55*m,1,mm) =
[2;2;2;2;2;3;3;3;3;3;4;4;4;4;4;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7];
    elseif a0 == 5
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 6
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    elseif a0 == 7
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 8
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    elseif a0 == 9
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 10
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;0;0;0;0;0;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    elseif a0 == 11
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;0;0;0;0;0;7;7;7;7;7];
    elseif a0 == 12
     AS((55*(m-1)+1):55*m,1,mm) =
[1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;1;0;0;0;0;0;7;7;7;7;7;3;3;3;3;3;7;7;7;7;7;5;6;5;6;5;6;5;6;5;6;7;7;7;7;7;7;7;7;7];
    end
 end
end

% Generate reasonable values of joint angles, external load force vectors, and torque generated when driving screw;
for nn = 1:10
 for n = 1:length(AS)
   ii0(n,1,nn) = lognrnd(2.9,0.8); jj0(n,1,nn) = randn*10.9+2.8; kk0(n,1,nn) = randn*12.9-3.2; % Trunk angles
   ii1(n,1,nn) = lognrnd(3.8,0.6); jj1(n,1,nn) = trirnd(0,45,90); kk1(n,1,nn) = trirnd(-20,0,20); % Shoulder angles
   ii2(n,1,nn) = trirnd(0,90,135); jj2(n,1,nn) = trirnd(-90,0,30); kk2(n,1,nn) = trirnd(-20,0,20); % Elbow angles
   ii3(n,1,nn) = trirnd(-20,0,20); jj3(n,1,nn) = trirnd(-20,0,20); kk3(n,1,nn) = trirnd(-20,0,20); % Wrist angles
   wy(n,1,nn) = 0; wz(n,1,nn) = -lognrnd(3.0,1.2)/2; mtx(n,1,nn) = 0; mtz(n,1,nn) = 0;

  % trunk posture code 1 (neutral)
  ii01 = lognrnd(2.9,0.8);
  while ii01>20
    ii01 = lognrnd(2.9,0.8);
  end
```

```
jj01 = randn*10.9+2.8;
while abs(jj01)>20
  jj01 = randn*10.9+2.8;
end
kk01 = randn*12.9-3.2;
while abs(kk01)>20
  kk01 = randn*12.9-3.2;
end

% trunk posture code 2 (mild flexion)
ii02 = lognrnd(2.9,0.8);
while (ii02<=20)|(ii02>45)
  ii02 = lognrnd(2.9,0.8);
end
jj02 = jj01;
kk02 = kk01;

% trunk posture code 3 (severe flexion)
ii03 = lognrnd(2.9,0.8);
while (ii03<=45)|(ii03>80)
  ii03 = lognrnd(2.9,0.8);
end
jj03 = jj01;
kk03 = kk01;

% trunk posture code 4 (bend & twist)
ii04 = ii01;
jj041 = randn*10.9+2.8;
jj042 = randn*10.9+2.8;
kk041 = randn*12.9-3.2;
kk042 = randn*12.9-3.2;
PL = 0.8;
PPL = (rand<=PL);
if PPL == 0
  while (jj041<-35)|(jj041>-20)
     jj041 = randn*10.9+2.8;
  end
  while (kk041<20)|(kk041>45)
     kk041 = randn*12.9-3.2;
  end
elseif PPL == 1
  while (jj042<20)|(jj042>35)
     jj042 = randn*10.9+2.8;
  end
  while (kk042<-45)|(kk042>-20)
     kk042 = randn*12.9-3.2;
  end
end

% trunk posture code 5 (bend & twist & flexion)
ii05 = lognrnd(2.9,0.8);
while (ii05<=20)|(ii05>80)
  ii05 = lognrnd(2.9,0.8);
end
jj051 = jj041;
jj052 = jj042;
kk051 = kk041;
kk052 = kk042;
```

```
% shoulder posture code 1 (2 arms down)
ii11 = lognrnd(3.8,0.6);
while ii11>60
  ii11 = lognrnd(3.8,0.6);
end
ii21 = trirnd(0,90,135);
while ii21<ii11
  ii21 = trirnd(0,90,135);
end

% Shoulder posture code 2 (1 arm up)
ii121 = lognrnd(3.8,0.6);
ii122 = lognrnd(3.8,0.6);
ii221 = trirnd(0,90,135);
ii222 = trirnd(0,90,135);
PR = 0.72;
PPR = (rand<=PR);
if PPR == 0 % Left arm up
  ii121 = ii11; ii221 = ii21;
elseif PPR == 1 % Right arm up
  while (ii122<60)|(ii122>135)
     ii122 = lognrnd(3.8,0.6);
  end
  while ii222<ii122
     ii222 = trirnd(0,90,135);
  end
end

% Shoulder posture code 3 (2 arms up)
 ii13 = ii122; ii23 = ii222;

switch AS(n,1,nn)
 case 0
  ii0(n,1,nn) = 0; jj0(n,1,nn) = 0; kk0(n,1,nn) = 0;
  ii1(n,1,nn) = 0; jj1(n,1,nn) = 0; kk1(n,1,nn) = 0;
  ii2(n,1,nn) = 0; jj2(n,1,nn) = 0; kk2(n,1,nn) = 0;
  ii3(n,1,nn) = 0; jj3(n,1,nn) = 0; kk3(n,1,nn) = 0;
  wx(n,1,nn) = 0; wy(n,1,nn) = 0; wz(n,1,nn) = 0;
  mtx(n,1,nn) = 0; mty(n,1,nn) = 0; mtz(n,1,nn) = 0;

 case 1; % activity 1 (cut/measure)
  wx(n,1,nn) = 10; wz(n,1,nn) = -2; mty(n,1,nn) = 0;

  c01 = simdiscr(1,[12,4,3,1,5]/25);
  if c01 == 1 % trunk posture code 1 (neutral)
    ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
  elseif c01 == 2 % trunk posture code 2 (mild flexion)
    ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
  elseif c01 == 3 % trunk posture code 3 (severe flexion)
    ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
  elseif c01 == 4 % trunk posture code 4 (bend & twist)
    ii0(n,1,nn) = ii04;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
    elseif PPL == 1
```

```
    jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
  end
 elseif c01 == 5 % trunk posture code 5 (bend & twist & flexion)
  ii0(n,1,nn) = ii05;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
  elseif PPL == 1
    jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
  end
 end

 c11 = simdiscr(1,[13,4,8]/25);
 if c11 == 1 % Shoulder posture code 1 (2 arms down)
  ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
 elseif c11 == 2 % Shoulder posture code 2 (1 arm up)
  PR = 0.72;
  PPR = (rand<=PR);
  if PPR == 0 % Left arm up
   ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
  elseif PPR == 1 % Right arm up
   ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
  end
 elseif c11 == 3 % Shoulder posture code 3 (2 arms up)
  ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
 end

case 2 % activity 2 (lift)
 kk3(n,1,nn) = -90;
 wx(n,1,nn) = 0; wz(n,1,nn) = -204; mty(n,1,nn) = 0;

 c02 = simdiscr(1,[0,2,1,0,5]/8);
 if c02 == 1 % trunk posture code 1 (neutral)
  ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
 elseif c02 == 2 % trunk posture code 2 (mild flexion)
  ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
 elseif c02 == 3 % trunk posture code 3 (severe flexion)
  ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
 elseif c02 == 4 % trunk posture code 4 (bend & twist)
  ii0(n,1,nn) = ii04;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
  elseif PPL == 1
    jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
  end
 elseif c02 == 5 % trunk posture code 5 (bend & twist & flexion)
  ii0(n,1,nn) = ii05;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
  elseif PPL == 1
    jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
  end
 end
```

```
c12 = simdiscr(1,[4,1,3]/8);
if c12 == 1 % Shoulder posture code 1 (2 arms down)
  ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
elseif c12 == 2 % Shoulder posture code 2 (1 arm up)
  PR = 0.72;
  PPR = (rand<=PR);
  if PPR == 0 % Left arm up
    ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
  elseif PPR == 1 % Right arm up
    ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
  end
elseif c12 == 3 % Shoulder posture code 3 (2 arms up)
  ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
end

case 3 % activity 3 (carry)
 kk3(n,1,nn) = -90;
 wx(n,1,nn) = 0; mty(n,1,nn) = 0;
 if (AS(n-1,1,nn) == 2)|(AS(n-2,1,nn) == 2)|(AS(n-3,1,nn) == 2)|(AS(n-4,1,nn) == 2)|(AS(n-5,1,nn) == 2)
    wz(n,1,nn) = -204;
 else
    wz1 = -lognrnd(3.0,1.2)/2;
     while wz1<-23/2
       wz1 = -lognrnd(3.0,1.2)/2;
     end
    wz2 = -lognrnd(3.0,1.2)/2;
     while (wz2<-67/2)|(wz2>-23/2)
       wz2 = -lognrnd(3.0,1.2)/2;
     end
    wz3 = -lognrnd(3.0,1.2)/2;
     while (wz3<-223/2)|(wz3>-67/2)
       wz3 = -lognrnd(3.0,1.2)/2;
     end
    wz4 = -lognrnd(3.0,1.2)/2;
     while (wz4<-408/2)|(wz4>-223/2)
       wz4 = -lognrnd(3.0,1.2/2);
     end
    c3wz = simdiscr(1,[1,5,4,6]/16);
    if c3wz == 1
      wz(n,1,nn) = wz1;
    elseif c3wz == 2
      wz(n,1,nn) = wz2;
    elseif c3wz == 3
      wz(n,1,nn) = wz3;
    elseif c3wz == 4
      wz(n,1,nn) = wz4;
    end
 end

 c03 = simdiscr(1,[15,0,1,0,0]/16);
 if c03 == 1 % trunk posture code 1 (neutral)
   ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
 elseif c03 == 2 % trunk posture code 2 (mild flexion)
   ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
 elseif c03 == 3 % trunk posture code 3 (severe flexion)
   ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
 elseif c03 == 4 % trunk posture code 4 (bend & twist)
```

```
  ii0(n,1,nn) = ii04;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
  elseif PPL == 1
    jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
  end
 elseif c03 == 5 % trunk posture code 5 (bend & twist & flexion)
  ii0(n,1,nn) = ii05;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
  elseif PPL == 1
    jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
  end
 end

 c13 = simdiscr(1,[10,3,3]/16);
 if c13 == 1 % Shoulder posture code 1 (2 arms down)
  ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
 elseif c13 == 2 % Shoulder posture code 2 (1 arm up)
  PR = 0.72;
  PPR = (rand<=PR);
  if PPR == 0 % Left arm up
    ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
  elseif PPR == 1 % Right arm up
    ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
  end
 elseif c13 == 3 % Shoulder posture code 3 (2 arms up)
  ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
 end

case 4 % activity 4 (hold/place)
 wx(n,1,nn) = 0; wz(n,1,nn) = -204; mty(n,1,nn) = 0;

 c04 = simdiscr(1,[0.33,0.17,0.33,0.17,0]);
 if c04 == 1 % trunk posture code 1 (neutral)
  ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
 elseif c04 == 2 % trunk posture code 2 (mild flexion)
  ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
 elseif c04 == 3 % trunk posture code 3 (severe flexion)
  ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
 elseif c04 == 4 % trunk posture code 4 (bend & twist)
  ii0(n,1,nn) = ii04;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
    jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
  elseif PPL == 1
    jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
  end
 elseif c04 == 5 % trunk posture code 5 (bend & twist & flexion)
  ii0(n,1,nn) = ii05;
  PL = 0.8;
  PPL = (rand<=PL);
  if PPL == 0
```

```
    jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
   elseif PPL == 1
    jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
   end
  end

  c14 = simdiscr(1,[3,0,3]/6);
  if c14 == 1 % Shoulder posture code 1 (2 arms down)
   ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
  elseif c14 == 2 % Shoulder posture code 2 (1 arm up)
   PR = 0.72;
   PPR = (rand<=PR);
   if PPR == 0 % Left arm up
    ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
   elseif PPR == 1 % Right arm up
    ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
   end
  elseif c14 == 3 % Shoulder posture code 3 (2 arms up)
   ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
  end

 case 5 % activity 5 (screw)
  jj1(n,1,nn) = 0; jj2(n,1,nn) = 0; jj3(n,1,nn) = 0;
  wx(n,1,nn) = 0; wy(n,1,nn) = -10; wz(n,1,nn) = -12; mty(n,1,nn) = -7;

  c05 = simdiscr(1,[7,5,3,3,2]/20);
  if c05 == 1 % trunk posture code 1 (neutral)
   ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
  elseif c05 == 2 % trunk posture code 2 (mild flexion)
   ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
  elseif c05 == 3 % trunk posture code 3 (severe flexion)
   ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
  elseif c05 == 4 % trunk posture code 4 (bend & twist)
   ii0(n,1,nn) = ii04;
   PL = 0.8;
   PPL = (rand<=PL);
   if PPL == 0
    jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
   elseif PPL == 1
    jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
   end
  elseif c05 == 5 % trunk posture code 5 (bend & twist & flexion)
   ii0(n,1,nn) = ii05;
   PL = 0.8;
   PPL = (rand<=PL);
   if PPL == 0
    jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
   elseif PPL == 1
    jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
   end
  end

  c15 = simdiscr(1,[9,6,5]/20);
  if c15 == 1 % Shoulder posture code 1 (2 arms down)
   ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
  elseif c15 == 2 % Shoulder posture code 2 (1 arm up)
   PR = 0.72;
   PPR = (rand<=PR);
```

```
    if PPR == 0 % Left arm up
      ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
    elseif PPR == 1 % Right arm up
      ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
    end
  elseif c15 == 3 % Shoulder posture code 3 (2 arms up)
    ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
  end

case 6 % activity 6 (in between)
  while jj1(n,1,nn)>45
    jj1(n,1,nn) = trirnd(0,45,90);
  end
  while jj2(n,1,nn)>jj1(n,1,nn)
    jj2(n,1,nn) = trirnd(-90,0,30);
  end
  wx(n,1,nn) = 0; wz(n,1,nn) = -12; mty(n,1,nn) = 0;

  c06 = simdiscr(1,[22,2,0,0,0]/24);
  if c06 == 1 % trunk posture code 1 (neutral)
    ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
  elseif c06 == 2 % trunk posture code 2 (mild flexion)
    ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
  elseif c06 == 3 % trunk posture code 3 (severe flexion)
    ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
  elseif c06 == 4 % trunk posture code 4 (bend & twist)
    ii0(n,1,nn) = ii04;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
    elseif PPL == 1
      jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
    end
  elseif c06 == 5 % trunk posture code 5 (bend & twist & flexion)
    ii0(n,1,nn) = ii05;
    PL = 0.8;
    PPL = (rand<=PL);
    if PPL == 0
      jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
    elseif PPL == 1
      jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
    end
  end

  c16 = simdiscr(1,[23,0,1]/24);
  if c16 == 1 % Shoulder posture code 1 (2 arms down)
    ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
  elseif c16 == 2 % Shoulder posture code 2 (1 arm up)
    PR = 0.72;
    PPR = (rand<=PR);
    if PPR == 0 % Left arm up
      ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
    elseif PPR == 1 % Right arm up
      ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
    end
  elseif c16 == 3 % Shoulder posture code 3 (2 arms up)
    ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
```

```
    end

  case 7 % activity 7 (other)
    wx(n,1,nn) = 0; mty(n,1,nn) = 0;
    while wz(n,1,nn)<-23
       wz(n,1,nn) = -lognrnd(3.0,1.2);
    end

    c07 = simdiscr(1,[13,2,8,0,4]/27);
    if c07 == 1 % trunk posture code 1 (neutral)
      ii0(n,1,nn) = ii01; jj0(n,1,nn) = jj01; kk0(n,1,nn) = kk01;
    elseif c07 == 2 % trunk posture code 2 (mild flexion)
      ii0(n,1,nn) = ii02; jj0(n,1,nn) = jj02; kk0(n,1,nn) = kk02;
    elseif c07 == 3 % trunk posture code 3 (severe flexion)
      ii0(n,1,nn) = ii03; jj0(n,1,nn) = jj03; kk0(n,1,nn) = kk03;
    elseif c07 == 4 % trunk posture code 4 (bend & twist)
      ii0(n,1,nn) = ii04;
      PL = 0.8;
      PPL = (rand<=PL);
      if PPL == 0
        jj0(n,1,nn) = jj041; kk0(n,1,nn) = kk041;
      elseif PPL == 1
        jj0(n,1,nn) = jj042; kk0(n,1,nn) = kk042;
      end
    elseif c07 == 5 % trunk posture code 5 (bend & twist & flexion)
      ii0(n,1,nn) = ii05;
      PL = 0.8;
      PPL = (rand<=PL);
      if PPL == 0
        jj0(n,1,nn) = jj051; kk0(n,1,nn) = kk051;
      elseif PPL == 1
        jj0(n,1,nn) = jj052; kk0(n,1,nn) = kk052;
      end
    end

    c17 = simdiscr(1,[18,2,7]/27);
    if c17 == 1 % Shoulder posture code 1 (2 arms down)
      ii1(n,1,nn) = ii11; ii2(n,1,nn) = ii21;
    elseif c17 == 2 % Shoulder posture code 2 (1 arm up)
      PR = 0.72;
      PPR = (rand<=PR);
      if PPR == 0 % Left arm up
        ii1(n,1,nn) = ii121; ii2(n,1,nn) = ii221;
      elseif PPR == 1 % Right arm up
        ii1(n,1,nn) = ii122; ii2(n,1,nn) = ii222;
      end
    elseif c17 == 3 % Shoulder posture code 3 (2 arms up)
      ii1(n,1,nn) = ii13; ii2(n,1,nn) = ii23;
    end
  end
 end
 U(:,:,nn) = [AS(:,nn) ii0(:,nn) jj0(:,nn) kk0(:,nn) ii1(:,nn) jj1(:,nn) kk1(:,nn) ii2(:,nn) jj2(:,nn) kk2(:,nn) ii3(:,nn) jj3(:,nn)
kk3(:,nn) wx(:,nn) wy(:,nn) wz(:,nn) mtx(:,nn) mty(:,nn) mtz(:,nn)];
 ANT(:,:,nn) = [ii0(:,nn) jj0(:,nn) kk0(:,nn)];
 ANS(:,:,nn) = [ii1(:,nn) jj1(:,nn) kk1(:,nn)];
 ANE(:,:,nn) = [ii2(:,nn) jj2(:,nn) kk2(:,nn)];
 ANW(:,:,nn) = [ii3(:,nn) jj3(:,nn) kk3(:,nn)];
 FEL(:,:,nn) = [wx(:,nn) wy(:,nn) wz(:,nn)];
```

```
 MT(:,:,nn) = [mtx(:,nn) mty(:,nn) mtz(:,nn)];
end

% Determine the center of mass position vectors for upper arm, forearm, and hand in the GH coordinate system;
for qq = 1:10
   H(qq) = randn*9.1+175.9;
   while (H(qq)<160)|(H(qq)>200)
     H(qq) = randn*9.1+175.9;
   end
   HEIGHT(qq) = H(qq)/100;
   CUA = 0.186; CFA = 0.146;
   S = [0; 0; 0]; UAS(:,qq) = [0; 0; -0.132/1.77*HEIGHT(qq)];
   E(:,qq) = [0; 0; -HEIGHT(qq)*CUA]; FAE(:,qq) = [0; 0; -0.117/1.77*HEIGHT(qq)];
   W(:,qq) = [0; 0; -HEIGHT(qq)*CFA]; HW(:,qq) = [0; 0; -0.07/1.77*HEIGHT(qq)];

   Weight(qq) = randn*13.3+79.8;
   while (Weight(qq)<65)|(Weight(qq)>110)
     Weight(qq) = randn*13.3+79.8;
   end
   WEIGHT(qq) = Weight(qq)*9.8; FUA(:,qq) = [0; 0; -21/737*WEIGHT(qq)]; FFA(:,qq) = [0; 0; -12/737*WEIGHT(qq)];
FH(:,qq) = [0; 0; -4/737*WEIGHT(qq)];
   FC(:,qq) = [0; 0; -1.5/737*WEIGHT(qq)]; FS(:,qq) = [0; 0; -7/737*WEIGHT(qq)];
end

% Compute the external force vectors and the external moment vectors in the GH and SC coordinate systems;
for rr = 1:10
   uas = UAS(:,rr); e = E(:,rr); fae = FAE(:,rr); w = W(:,rr); hw = HW(:,rr);
   fua = FUA(:,rr); ffa = FFA(:,rr); fh = FH(:,rr); fc = FC(:,rr); fs = FS(:,rr);
  for r = 1:length(AS)
   ii0 = ANT(r,1,rr); jj0 = ANT(r,2,rr); kk0 = ANT(r,3,rr);
   ii1 = ANS(r,1,rr); jj1 = ANS(r,2,rr); kk1 = ANS(r,3,rr);
   ii2 = ANE(r,1,rr); jj2 = ANE(r,2,rr); kk2 = ANE(r,3,rr);
   ii3 = ANW(r,1,rr); jj3 = ANW(r,2,rr); kk3 = ANW(r,3,rr);
   fel = FEL(r,:,rr)'; t = MT(r,:,rr)';
   % Obtain upper extremity rotation matrix (T0) based on trunk motion;
   i0 = ii0*pi/180; j0 = jj0*pi/180; k0 = kk0*pi/180;
   T0 = [      cos(j0)*cos(k0)                    cos(j0)*sin(k0)                   -sin(j0)
      -cos(i0)*sin(k0)+sin(i0)*sin(j0)*cos(k0)   cos(i0)*cos(k0)+sin(i0)*sin(j0)*sin(k0)   sin(i0)*cos(j0)
       sin(i0)*sin(k0)+cos(i0)*sin(j0)*cos(k0)  -sin(i0)*cos(k0)+cos(i0)*sin(j0)*sin(k0)   cos(i0)*cos(j0)];
   % Obtain glenohumeral joint rotation matrix (T1);
   i1 = -ii1*pi/180; j1 = jj1*pi/180; k1 = kk1*pi/180;
   T1 = [      cos(j1)*cos(k1)                    cos(j1)*sin(k1)                   -sin(j1)
      -cos(i1)*sin(k1)+sin(i1)*sin(j1)*cos(k1)   cos(i1)*cos(k1)+sin(i1)*sin(j1)*sin(k1)   sin(i1)*cos(j1)
       sin(i1)*sin(k1)+cos(i1)*sin(j1)*cos(k1)  -sin(i1)*cos(k1)+cos(i1)*sin(j1)*sin(k1)   cos(i1)*cos(j1)];
   % Obtain elbow joint rotation matrix (T2);
   i2 = -(ii2-ii1)*pi/180; j2 = jj2*pi/180; k2 = kk2*pi/180;
   T2 = [      cos(j2)*cos(k2)                    cos(j2)*sin(k2)                   -sin(j2)
      -cos(i2)*sin(k2)+sin(i2)*sin(j2)*cos(k2)   cos(i2)*cos(k2)+sin(i2)*sin(j2)*sin(k2)   sin(i2)*cos(j2)
       sin(i2)*sin(k2)+cos(i2)*sin(j2)*cos(k2)  -sin(i2)*cos(k2)+cos(i2)*sin(j2)*sin(k2)   cos(i2)*cos(j2)];
   % Obtain wrist joint rotation matrix (T3);
   i3 = ii3*pi/180; j3 = jj3*pi/180; k3 = kk3*pi/180;
   T3 = [      cos(j3)*cos(k3)                    cos(j3)*sin(k3)                   -sin(j3)
      -cos(i3)*sin(k3)+sin(i3)*sin(j3)*cos(k3)   cos(i3)*cos(k3)+sin(i3)*sin(j3)*sin(k3)   sin(i3)*cos(j3)
       sin(i3)*sin(k3)+cos(i3)*sin(j3)*cos(k3)  -sin(i3)*cos(k3)+cos(i3)*sin(j3)*sin(k3)   cos(i3)*cos(j3)];

   ua = T0*(T1*uas); fa = T0*(T1*(T2*fae+e)); h = T0*(T1*(T2*(T3*hw+w)+e));
   fegh = fua+ffa+fh+fel; megh = cross(ua,fua)+cross(fa,ffa)+cross(h,fh+fel)+t;
   fesc = fua+ffa+fh+fel+fc+fs;
```

```
   FEGH(:,r,rr) = fegh; B(:,r,rr) = -megh;
   FESC(:,r,rr) = fesc;
  end
end

% Obtain muscle force vector F and muscle-force-generated moment vector M;
for ss = 1:10
 for s = 1:length(AS)
   % Determine the Euler angles of the clavicula system based on the sternum system;
   xx1(s,1,ss) = randn*10.9+323.0;
   while (xx1(s,1,ss)<300.0)|(xx1(s,1,ss)>348.2)
    xx1(s,1,ss) = randn*10.9+323.0;
   end
   yy1(s,1,ss) = randn*8.2+24.8;
   while (yy1(s,1,ss)<10.0)|(yy1(s,1,ss)>47.2)
    yy1(s,1,ss) = randn*8.2+24.8;
   end
   zz1(s,1,ss) = randn*7.6+8.5;
   while (zz1(s,1,ss)<0.0)|(zz1(s,1,ss)>27.5)
    zz1(s,1,ss) = randn*7.6+8.5;
   end
   % Determine the Euler angles of the scapula system based on the sternum system;
   xx2(s,1,ss) = randn*33.8+191.8;
   while (xx2(s,1,ss)<200.0)|(xx2(s,1,ss)>253.3)
    xx2(s,1,ss) = randn*33.8+191.8;
   end
   yy2(s,1,ss) = randn*1.8+8.6;
   while (yy2(s,1,ss)<-74.7)|(yy2(s,1,ss)>0.0)
    yy2(s,1,ss) = randn*1.8+8.6;
   end
   zz2(s,1,ss) = randn*21.8+82.0;
   while (zz2(s,1,ss)<90.0)|(zz2(s,1,ss)>122.9)
    zz2(s,1,ss) = randn*21.8+82.0;
   end
   % Determine the Euler angles of the humerus system based on the sternum system;
   xx3(s,1,ss) = randn*3.5+38.1;
   while (xx3(s,1,ss)<0.0)|(xx3(s,1,ss)>80.9)
    xx3(s,1,ss) = randn*3.5+38.1;
   end
   yy3(s,1,ss) = randn*9.2-2.5;
   while (yy3(s,1,ss)<-34.6)|(yy3(s,1,ss)>0.0)
    yy3(s,1,ss) = randn*9.2-2.5;
   end
   zz3(s,1,ss) = randn*20.3+77.1;
   while (zz3(s,1,ss)<41.3)|(zz3(s,1,ss)>90.0)
    zz3(s,1,ss) = randn*20.3+77.1;
   end

   % Compute the rotation matrices of the clavicula, scapula, and humerus systems based on the sternum system;
   x1 = xx1(s,1,ss)*pi/180; y1 = -yy1(s,1,ss)*pi/180; z1 = zz1(s,1,ss)*pi/180;
   R12 = [      cos(y1)*cos(z1)                  -cos(y1)*sin(z1)              sin(y1)
       cos(x1)*sin(z1)+sin(x1)*sin(y1)*cos(z1)  cos(x1)*cos(z1)-sin(x1)*sin(y1)*sin(z1)  -sin(x1)*cos(y1)
       sin(x1)*sin(z1)-cos(x1)*sin(y1)*cos(z1)  sin(x1)*cos(z1)+cos(x1)*sin(y1)*sin(z1)  cos(x1)*cos(y1)];
   x2 = xx2(s,1,ss)*pi/180; y2 = -yy2(s,1,ss)*pi/180; z2 = zz2(s,1,ss)*pi/180;
   R22 = [      cos(y2)*cos(z2)                  -cos(y2)*sin(z2)              sin(y2)
       cos(x2)*sin(z2)+sin(x2)*sin(y2)*cos(z2)  cos(x2)*cos(z2)-sin(x2)*sin(y2)*sin(z2)  -sin(x2)*cos(y2)
       sin(x2)*sin(z2)-cos(x2)*sin(y2)*cos(z2)  sin(x2)*cos(z2)+cos(x2)*sin(y2)*sin(z2)  cos(x2)*cos(y2)];
   x3 = xx3(s,1,ss)*pi/180; y3 = -yy3(s,1,ss)*pi/180; z3 = zz3(s,1,ss)*pi/180;
```

```
R32 = [       cos(y3)*cos(z3)                    -cos(y3)*sin(z3)              sin(y3)
    cos(x3)*sin(z3)+sin(x3)*sin(y3)*cos(z3)  cos(x3)*cos(z3)-sin(x3)*sin(y3)*sin(z3)  -sin(x3)*cos(y3)
    sin(x3)*sin(z3)-cos(x3)*sin(y3)*cos(z3)  sin(x3)*cos(z3)+cos(x3)*sin(y3)*sin(z3)  cos(x3)*cos(y3)];

% Determine the relationship among the origins of different coordinate systems;
X = 0.141; Y = -0.057; Z = -0.01;
T = [X;Y;Z]; % GH @ SC
XX = 0.112; YY = -0.058; ZZ = 0.025;
TT = [XX;YY;ZZ]; % AC @ SC
XXX = 0.029; YYY = 0.002; ZZZ = -0.035;
TTT = [XXX;YYY;ZZZ]; % GH @ AC

% determine the complement of the elbow angle
ii1 = ANS(s,1,ss); ii2 = ANE(s,1,ss);
anes = ii2-ii1;
if anes>90
   ance = (ii2-ii1)-90;
else
   ance = 0;
end
AN = ance*pi/180;

% Calculate the muscle unit force vectors and muscle moment vectors in the GH coordinate system;
jj1 = ANS(s,2,ss); kk1 = ANS(s,3,ss);
i1 = -ii1*pi/180; j1 = jj1*pi/180; k1 = kk1*pi/180;
T11 = [cos(k1)*cos(j1)-sin(k1)*cos(i1)*sin(j1)   cos(k1)*sin(j1)+sin(k1)*cos(i1)*cos(j1)  sin(k1)*sin(i1)
    -sin(k1)*cos(j1)-cos(k1)*cos(i1)*sin(j1)  -sin(k1)*sin(j1)+cos(k1)*cos(i1)*cos(j1)  cos(k1)*sin(i1)
          sin(i1)*sin(j1)                  -sin(i1)*cos(j1)                cos(i1)   ];

a1 = T11*R32*[0.170;0.049;0.017];                    b1 = T11*([0.260;0;-0.750]-T);
a2 = T11*R32*[0.170;0.049;0.017];                    b2 = T11*([0.298;-0.266;-0.809]-T);
a3 = T11*R32*[0.275;0.040;0.043];                    b3 = T11*([0;0.148;-0.209]-T);
a4 = T11*R32*[0.275;0.040;0.043];                    b4 = T11*(R12*[0.374;0.097;0.027]-T);
a5 = T11*R32*[0.369;-0.004;0.044];                   b5 = T11*(R22*[-0.009;-0.147;-0.062]-TTT);
a6 = T11*R32*[0.369;-0.004;0.044];                   b6 = T11*(R22*[0.241;0.199;-0.174]-TTT);
a7 = T11*R32*[0.369;-0.004;0.044];                   b7 = T11*(R12*[0.795;-0.028;0.015]-T);
a8 = T11*R32*[0.396;0.014;-0.018];                   b8 = T11*(R22*[0.118;-0.014;0.193]-TTT);
a9 = T11*R32*[-0.018;-0.025;0.094];                  b9 = T11*(R22*[0.617;0.240;-0.106]-TTT);
a10 = T11*R32*[-0.018;-0.025;0.094];                 b10 = T11*(R22*[0.760;0.115;-0.069]-TTT);
a11 = T11*R32*[0.012;0.085;0.011];                   b11 = T11*(R22*[0.523;0.307;-0.137]-TTT);
a12 = T11*R32*[0.012;0.085;0.011];                   b12 = T11*(R22*[0.668;0.164;-0.108]-TTT);
a13 = T11*R32*[0.012;0.085;0.011];                   b13 = T11*(R22*[0.762;0.069;-0.064]-TTT);
a14 = T11*R32*[-0.042;0.031;0.079];                  b14 = T11*(R22*[0.305;0.348;-0.112]-TTT);
a15 = T11*R32*[0.143;0.037;-0.022];                  b15 = T11*(R22*[0.844;-0.058;-0.006]-TTT);
a16 = T11*R32*[0.039;-0.045;0.037];                  b16 = T11*(R22*[0.600;-0.049;-0.024]-TTT);
a17 = T11*R32*[1.036+0.143*cos(AN);0.143*sin(AN);0];       b17 = T11*(R22*[0.102;0.016;0.014]-TTT);
a18 = T11*R32*[1.036+0.143*cos(AN);0.143*sin(AN);0];       b18 = T11*(R22*[0.051;0.007;0.227]-TTT);
a19 = T11*R32*[1.036-0.64*sin(0.73-AN);-0.64*cos(0.73-AN);0]; b19 = T11*(R22*[0.366;-0.109;0.041]-TTT);
a20 = T11*R32*[-0.035;0.050;0.053];                  b20 = T11*(R22*[0.080;0.080;0.260]-TTT);

F1 = (a1-b1)/norm(a1-b1); M1 = cross(a1,F1);
F2 = (a2-b2)/norm(a2-b2); M2 = cross(a2,F2);
F3 = (a3-b3)/norm(a3-b3); M3 = cross(a3,F3);
F4 = (a4-b4)/norm(a4-b4); M4 = cross(a4,F4);
F5 = (a5-b5)/norm(a5-b5); M5 = cross(a5,F5);
F6 = (a6-b6)/norm(a6-b6); M6 = cross(a6,F6);
F7 = (a7-b7)/norm(a7-b7); M7 = cross(a7,F7);
F8 = (a8-b8)/norm(a8-b8); M8 = cross(a8,F8);
```

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F9 = (a9-b9)/norm(a9-b9); M9 = cross(a9,F9);
F10 = (a10-b10)/norm(a10-b10); M10 = cross(a10,F10);
F11 = (a11-b11)/norm(a11-b11); M11 = cross(a11,F11);
F12 = (a12-b12)/norm(a12-b12); M12 = cross(a12,F12);
F13 = (a13-b13)/norm(a13-b13); M13 = cross(a13,F13);
F14 = (a14-b14)/norm(a14-b14); M14 = cross(a14,F14);
F15 = (a15-b15)/norm(a15-b15); M15 = cross(a15,F15);
F16 = (a16-b16)/norm(a16-b16); M16 = cross(a16,F16);
F17 = (a17-b17)/norm(a17-b17); M17 = cross(a17,F17);
F18 = (a18-b18)/norm(a18-b18); M18 = cross(a18,F18);
F19 = (a19-b19)/norm(a19-b19); M19 = cross(a19,F19);
F20 = (a20-b20)/norm(a20-b20); M20 = cross(a20,F20);

FGH(:,:,s,ss) = [F1 F2 F3 F4 F5 F6 F7 F8 F9 F10 F11 F12 F13 F14 F15 F16 F17 F18 F19 F20];
MGH(:,:,s,ss) = [M1 M2 M3 M4 M5 M6 M7 M8 M9 M10 M11 M12 M13 M14 M15 M16 M17 M18 M19 M20];

% Calculate the muscle unit force vectors and muscle moment vectors in the SC coordinate system;
A1 = a1+T; B1 = [0.260;0;-0.750];
A2 = a2+T; B2 = [0.298;-0.266;-0.809];
A3 = a3+T; B3 = [0;0.148;-0.209];
A4 = a4+T; B4 = R12*[0.374;0.097;0.027];
A5 = a5+T; B5 = R22*[-0.009;-0.147;-0.062]+TT;
A6 = a6+T; B6 = R22*[0.241;0.199;-0.174]+TT;
A7 = a7+T; B7 = R12*[0.795;-0.028;0.015];
A8 = a8+T; B8 = R22*[0.118;-0.014;0.193]+TT;
A9 = a9+T; B9 = R22*[0.617;0.240;-0.106]+TT;
A10 = a10+T; B10 = R22*[0.760;0.115;-0.069]+TT;
A11 = a11+T; B11 = R22*[0.523;0.307;-0.137]+TT;
A12 = a12+T; B12 = R22*[0.668;0.164;-0.108]+TT;
A13 = a13+T; B13 = R22*[0.762;0.069;-0.064]+TT;
A14 = a14+T; B14 = R22*[0.305;0.348;-0.112]+TT;
A15 = a15+T; B15 = R22*[0.844;-0.058;-0.006]+TT;
A16 = a16+T; B16 = R22*[0.600;-0.049;-0.024]+TT;
A17 = a17+T; B17 = R22*[0.102;0.016;0.014]+TT;
A18 = a18+T; B18 = R22*[0.051;0.007;0.227]+TT;
A19 = a19+T; B19 = R22*[0.366;-0.109;0.041]+TT;
A20 = a20+T; B20 = R22*[0.080;0.080;0.260]+TT;

F11 = (B1-A1)/norm(B1-A1); M11 = cross(B1, F11);
F21 = (B2-A2)/norm(B2-A2); M21 = cross(B2, F21);
F31 = (B3-A3)/norm(B3-A3); M31 = cross(B3, F31);
F41 = (B4-A4)/norm(B4-A4); M41 = cross(B4, F41);
F51 = (B5-A5)/norm(B5-A5); M51 = cross(B5, F51);
F61 = (B6-A6)/norm(B6-A6); M61 = cross(B6, F61);
F71 = (B7-A7)/norm(B7-A7); M71 = cross(B7, F71);
F81 = (B8-A8)/norm(B8-A8); M81 = cross(B8, F81);
F91 = (B9-A9)/norm(B9-A9); M91 = cross(B9, F91);
F101 = (B10-A10)/norm(B10-A10); M101 = cross(B10, F101);
F111 = (B11-A11)/norm(B11-A11); M111 = cross(B11, F111);
F121 = (B12-A12)/norm(B12-A12); M121 = cross(B12, F121);
F131 = (B13-A13)/norm(B13-A13); M131 = cross(B13, F131);
F141 = (B14-A14)/norm(B14-A14); M141 = cross(B14, F141);
F151 = (B15-A15)/norm(B15-A15); M151 = cross(B15, F151);
F161 = (B16-A16)/norm(B16-A16); M161 = cross(B16, F161);
F171 = (B17-A17)/norm(B17-A17); M171 = cross(B17, F171);
F181 = (B18-A18)/norm(B18-A18); M181 = cross(B18, F181);
F191 = (B19-A19)/norm(B19-A19); M191 = cross(B19, F191);
F201 = (B20-A20)/norm(B20-A20); M201 = cross(B20, F201);
```

```
 FSC(:,:,s,ss) = [F11 F21 F31 F41 F51 F61 F71 F81 F91 F101 F111 F121 F131 F141 F151 F161 F171 F181 F191
F201];
 MSC(:,:,s,ss) = [M11 M21 M31 M41 M51 M61 M71 M81 M91 M101 M111 M121 M131 M141 M151 M161 M171
M181 M191 M201];
 end
end

% Run the nonlinear optimization program to obtain muscle forces and the GH and SC joint resultant forces.
x0 = [0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0];
LBnd = [0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0;0];
co = 70; PCSA = [2.61;3.15;4.50;2.75;4.29;5.63;5.21;1.91;3.87;4.05;3.13;2.82;3.87;3.83;4.48;1.89;2.14;1.89;8.87];
UBnd = [co*PCSA;10000];
options = optimset('MaxFunEvals',100000,'MaxIter',100000);
for uu = 1:10
  for u = 1:length(AS)
    Aeq = MGH(:,:,u,uu); beq = B(:,u,uu);
    fgh = FGH(:,:,u,uu);
    fsc = FSC(:,:,u,uu);
    fegh1 = FEGH(:,u,uu);
    fesc1 = FESC(:,u,uu);
    uuu = U(u,:,uu);
    [x,f,exitflag] = fmincon('MuscleForceObjFunct',x0,[],[],Aeq,beq,LBnd,UBnd,[],options);
    ngh = -fgh*x-fegh1; fngh = norm(ngh);
    nsc = -fsc*x-fesc1; fnsc = norm(nsc);
    MF = [x;fngh;fnsc]';
    ANSWER(u,:,uu) = [uuu MF];
 end
end
```

Appendix D

MATLAB curve-fitting program to solve for trunk muscle fatigue parameters

```
% Obtain the trunk muscle recovery rate (m) and decay rate (n) using the curve fitting
optimization method.
a = [1/0.25,1/0.5,1/0.75];
b = [10.1,2.4,0.9];
x0 = [0.001,0.001];
MVCETFit = inline('-log(1+x(1)*(1-a))/(x(1)*x(2))','x','a');
[x,resid] = lsqcurvefit(MVCETFit,x0,a,b);
m = x(1)*x(2); n = x(2);
```

Biographical Sketch of Author

With a major of safety engineering, Lu Yuan obtained a B.E. degree from Beijing Institute of Technology, China in July, 2000. After graduation, he worked as a quality control engineer in International Flavors & Fragrances (China) Ltd., Guangzhou, China. The experience of watching people in the production lines develop adverse health problems from awkward working postures, exposure to toxic chemicals and low temperatures has motivated him to learn ergonomics worldwide. In 2001, he came to the University of Massachusetts Lowell where he received a M.S. degree in Occupational Ergonomics in May 2004 and is now on track to earn a doctoral degree in ergonomics. During the past five years, Lu has conducted scientific research in the multidisciplinary areas including construction, auto mechanics, and healthcare. Of note, he has worked as a research assistant for the Construction Occupational Health Program (COHP), a multi-year research effort dedicated in the promotion of occupational safety and health among construction workers. The opportunities of leading a number of student chapters and associations also add him the confidence of contributing to the scientific field of ergonomics in the future.