University of Alberta

Work-Related Low Back Disorders in Heavy Jobs and their Control

by

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Abstract

Work-related low back disorders (WLBD), the most frequent and costly musculoskeletal disorders, are related to working demands. Female nurses and male steel workers (SW) are among professionals with highest WLBD rates. This project studied WLBD in an acute-care hospital and in two steel companies. Injury records were reviewed and a questionnaire survey was conducted on 47 nurses and 108 SW. The working-life WLBD incidence rate and low back pain point prevalence were 65% and 30% for orthopedic nurses (ON), 58% and 25% for ICU nurses (IN), and 36% and 16% for SW. Mean(standard deviation) perceived exertion (10-point scale) was 7(2) for ON, 6(2) for IN, and 5(1) for SW. Patient transfers (ON), turning/repositioning patients in bed (IN), and punching holes into steel blades (SW) were considered the most demanding/risky parts of their occupation. A functional capacity evaluation (25 nurses and 25 SW) and a biomechanical demand analysis of manual handling and patient transfers (36 nurses), and of punching tasks (23 SW) were performed. The job simulated forces [79(16)% of the nurses maximum and 72(22)% of the SW maximum] were higher than the preferred levels [56(21)% and 55(17)% of the nurses and SW maximum, $p<0.01$]. Lumbar lordosis (mean difference=8°, $p<0.001$) and range of extension (mean difference=4°, $p=0.004$) and rotation (mean difference=5°, $p<0.001$) were higher for nurses than for SW. The instantaneous compression at L5/S1 [4754(437)N] and population without sufficient torso strength [37(9)%] were highest during the bed-to-stretcher transfers' pushing phase. The shear force [487(40)N] and ligament strain [14(5)%] were highest during the stretcher-to-bed transfers' pulling phase. Instantaneous compression during the positioning phase of the SW punching task was 2828(318)N and shear was 219(33)N. Between 95% and 99%

of males would be capable of performing this task. However, the lumbar flexion used was close to the full rage [90(13)%]. Evidence-based recommendations for modifications and training programs to reduce the risk of WLBD have been proposed. This combined methodology is useful in identifying specific risks for WLBD. Fitness for work, job modifications, and training programs can be designed based on the results.

Keywords: musculoskeletal disorders, low back, prevention, physical ergonomics

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List of Abbreviations

- Action Limit (AL)
- Actual Weight Lifted (AWL)
- Alberta Human Resources and Employment (AHRE)
- American Medical Association (AMA)
- American Physical Therapy Association (APTA)
- Angular Motion (AM)
- Biomechanical Model (BM)
- Body Mass Index (BMI)
- Bureau of Labor Statistics (BLS)
- Coefficient of Variation (CV)
- Computer Numeric Control Workers (CNC)
- Confidence Interval (CI)
- Cross-Sectional Area (CSA)
- Determination Coefficient (r^2)
- Electrogoniometers (elgons)
- Electromyography (EMG)
- Faculty of Graduate Studies and Research (FGSR)
- Functional Capacity Evaluation (FCE)
- ICU nurses (INs)
- International Ergonomics Association (IEA)
- Intra-Abdominal Pressure (IAP)
- Intra-Class Correlation Coefficient (ICC)
- Lifting Strength Rating (LSR)
- Lumbar Motion Monitor (LMM)
- Magnetic Resonance Imaging (MRI)
- Maximal Acceptable Weight (MAW)
- Maximum Isometric Strengths (MIS)
- Maximum Permissible Limit (MPL)
- Maximum Voluntary Contraction (MVC)

- Maximum Voluntary Exertion (MVE)
- Movement Demand Index (MDI)
- National Institute for Occupational Safety and Health (NIOSH)
- Occupational Safety and Health Administration (OSHA)
- Odds Ratio (OR)
- Orthopedic Nurses (ONs)
- Ovaco Working Postures Assessment System (OWAS)
- Pearson Correlation Coefficient (*r*)
- Physical Demands Analysis (PDA)
- Population (*N*)
- Probability (*p*)
- Range of Motion (ROM)
- Rapid Entire Body Assessment (REBA)
- Rapid Upper Limb Assessment (RULA)
- Recommended Weight Limit (RWL)
- Relative Risk (RR)
- Repositioning Up In Bed (U-B)
- Root Mean Square (RMS)
- Sample Size (*n*)
- Standard Deviation (SD)
- Steel Workers (SW)
- Three-Dimensional (3-D)
- Transferring from Bed to Stretcher (B-S)
- Transferring from Stretcher to Bed (S-B)
- Turning Away (T-A)
- Turning Towards (T-T)
- Two-Dimensional (2-D)
- Ultimate Compressive Strength (UCS)
- University of Alberta (UofA)
- Visual Analogue Scales (VAS)

- Visual Display Unit (VDU)
- Workers' Compensation Board (WCB)
- Work-Related Low Back Disorders (WLBD)
- Work-Related Musculoskeletal Disorders (WMSD)
- World Health Organization (WHO)

Chapter 1 – Introduction

1.1. Overview of the Thesis

I started my PhD in September 2002 after finishing my masters degree in Brazil (MSc PT 2000-2002). This thesis is a result of my doctoral study in Rehabilitation Science (Ergonomics Research Laboratory), Faculty of Rehabilitation Medicine, University of Alberta (UofA). The thesis is divided in fifteen chapters and was written in the paper format allowed by the Faculty of Graduate Studies and Research (FGSR, 2006). Chapters 1 to 5 represents the foundation of the research conducted, and chapters 6 to 15 present the results, discussion, and conclusions of the studies and thesis as a whole.

Chapter 1 is the introduction to the thesis and presents a literature review leading to the objectives of this research. The motivation for this work was to discover common ground between ergonomics science and consulting practice, and to advance the strategies and methods of work-related musculoskeletal disorders (WMSD) control and prevention. This research focused on work-related low back disorders (WLBD) in heavy jobs. It developed and used a combined methodology to study WLBD, and generated suggestions for interventions to control WLBD in jobs with risk factors. Two steel companies and one acute care teaching hospital in Edmonton, Alberta, Canada participated and supported the research which included an epidemiological and questionnaire study and biomechanical demands analysis to arrive at recommendations for modifications and for training programs to minimize the risk of WLBD. Thus, this research involved identifying, evaluating, and suggesting modifications of jobs with high incidence of WLBD.

Chapter 2 presents a literature review on working postures and devices to measure body kinematics. Chapter 3 presents a classification system to assess the biomechanical demands of different jobs. Chapter 4 presents a review of the biomechanical models used to estimate spinal loading in workplace settings, and Chapter 5 presents a review of cut-points for force exertion at work.

Chapter 6 presents the results of the epidemiological and questionnaire survey of nurses, and Chapter 7 presents the same information for the steel workers. Chapters 8 and 9 present the results of the force and electromyographic study of pushing, pulling, and lifting by the nurses and steel workers, respectively, and Chapter 10 compares the nurses and steel workers. Chapters 11 and 12 present the results of the lumbar motion and loading during patient handling by the nurses and the punching job task by the steel workers, respectively. Chapter 13 presents the results of the comparison of the lumbar range of motion and lordosis of the nurses and steel workers. Chapter 14 is a general discussion based on the results of all the research conducted presenting a framework for the disciplines of work physical therapy and rehabilitation ergonomics. This chapter integrates the results of the studies and presents suggestions of measures to control WLBD in the jobs analyzed. Finally, Chapter 15 presents the conclusions of the research conducted.

1.2. Work-Related Musculoskeletal Disorders (WMSD) and Work Biomechanical Demands

According to the U.S. Department of Labor, Bureau of Labor Statistics (BLS) "work-related injuries and illnesses are events or exposures in the work environment that caused or contributed to the condition or significantly aggravated a preexisting condition", and musculoskeletal disorders are "injuries or disorders of the muscles, nerves, tendons, joints, cartilage, or spinal discs" including sprains, strains, tears, soreness, pain, carpal tunnel syndrome, hernias, and connective tissue injuries that were caused by reactions of the body during work such as overexertion and repetition on bending, climbing, crawling, reaching, and twisting (BLS, 2001 and 2005). In the U.S. alone, more than 600 thousand workers have a lost time due to WMSD each year (United Electrical, Radio and Machine Workers of America, 1999). It was estimated that in 1995 in the U.S. the cost of WMSD was approximately $215 billion (Praemer *et al.*, 1999). The work biomechanical demands can cause and aggravate WMSD (Kumar 2001). According to the National Institute for Occupational Safety and Health (NIOSH), several epidemiological studies (review of over 700 studies) have demonstrated evidence of a causal relationship between physical exertion at work and WMSD (Bernard, 1997).

Westgaard and Winkel (1997) defined physical exertion (mechanical exposure) as "mechanical forces generated to meet work demands, considering level, repetitiveness and duration". Posture, motion, force, repetition, and duration must be considered in order to categorize the work biomechanical demands. These are important factors to be taken into account for musculoskeletal safety in the workplace because they are the mechanical variables related to morbidity (Armstrong, 1986; Coury, 1999; Kumar, 1994; Magnusson and Pope, 1998; Westgard and Winkel, 1997).

Biomechanical hazards, genetic predisposition, morphological disadvantages, and psychosocial propensity interact in the precipitation of WMSD (the Multivariate Interaction Theory – Kumar, 2001). Awkward, constrained, asymmetric, repeated, and prolonged postures; overstressing movements, high repetition; and force can overload the tissues and exceed their threshold of tolerable stress, causing injury due to overexertion or imbalance (the Overexertion Theory and the Differential Fatigue Theory – Kumar, 2001). Physical exertion is influenced by the task, workstation, design of working tools, and by the anthropometric characteristics of the workers. Maintenance of static exertion for prolonged periods of time compresses the veins and capillaries inside the muscles, causing micro lesions due to the absence of tissue oxygenation and nutrition. All these factors can cause imbalance, fatigue, discomfort, and pain due to disruption of tissues (Kumar 1990 and 2001).

1.3. Epidemiology and Costs of Work-Related Low Back Disorders (WLBD)

WLBD are the most common and represent the most costly WMSD and a great deal of suffering (AHRE, 2006; BLS, 2001; Nordin *et al.*, 1997). It is estimated that between 60% and 80% of the population will present with at least one episode of low back pain during their lifetime (e.g. Department of Education, 1993). In 1988, 7 million Americans were suffering from low back pain (Gill *et al.*, 1988). In 1993, more than five million people were out of work due to WLBD alone. Of these, two million were permanently injured, and for every ten adults, at least one presented with chronic back pain associated with physical incapacities such as difficulties in walking, standing upright, sitting and standing up.

An episode of WLBD causes fourteen lost working days on average (Department of Education, 1993); Andersson and Deyo (1997) studied the recovery time of 49,000 patients with WLBD, and found that 57% recovered in one week, 90% in six weeks, 95% in twelve weeks, and 1.2% were still work-disabled after 1 year. From 1985 to 2000, WLBD classified as strain, sprain, or contusion was the most common compensation claim in the Virginia Workers' Compensation Commission database (9.1%, N = 76,025); motion and overexertion were the mechanism of injury in 87% of the WLBD (Enders and Walker, 2003).

In 2000, based on the National Health Interview Survey, from 11 to 13 million people developed low back pain, and about $100 billion were spent on this problem in the U.S. alone. Considering the people who had worked in the previous year, 22.4 million had back pain (56% were males and 44% were females). The male workers with the highest risk (prevalence ratio > 2) were construction employees, carpenters, and industrial truck and tractor operators; while the highest risk female workers were nurses, orderlies, attendants, maids, janitors, and cleaners (Kumar, 2004; National Academy of Sciences, 2001). The incidence rate of lost time WLBD per 10,000 full-time American workers in 2001 was 20.2 (BLS, 2001). Sixty-five percent of all WLBD in 2001 in the U.S. were caused by overexertion – 60% of these occurred in lifting.

The occupational groups with the most WLBD cases (40%) were operators, fabricators, and industrial labors (BLS, 2001). From 2000 to 2005, between 26% and 27% of all lost time claims accepted by the Workers' Compensation Board of Alberta were related to WLBD; with the back being the body part most often injured (e.g. AHRE, 2006). Approximately 70% of the WLBD were sprains, strains and tears, and approximately 70% of these resulted from overexertion while pulling, pushing, lifting, carrying, twisting, climbing, tripping, and reaching.

1.4. Work-Related Low Back Disorders (WLBD): causes and dispute, definition and their control

1.4.1. Causes and dispute

Due to the multifactor determination of WMSD precipitation and the current dispute about the causes of WLBD, it is pertinent to discuss the role of different factors in the causation of WLBD. The source of WLBD may be explained by a biomechanical paradigm which assumes mechanical disruption of musculoskeletal tissues on the lumbar spine region, or by biochemical models which assume changes in the biochemical characteristics and balance, or by psychosocial variables which assume a predisposition to "reporting" WLBD due to personality, job satisfaction, or financial or social interests.

Overexertion resulting in injury by disruption of tissue can cause WLBD (Kumar, 1994). However, no "injured tissue" or objective indication of injury is expected to be found in many cases (e.g. Deyo and Weinstein, 2001; Frank *et al.*, 1996). Kumar (1994) defined overexertion as "a physical activity in which the level of effort exceeds normal physiological and mechanical (physical) tolerance limits". The current imaging techniques used as diagnostic tools, such as Magnetic Resonance Imaging (MRI), allow for visualization of different tissues and structures. However, injuries or the absence of visible injuries do not correlate with reported pain or disability in many cases. For example, studies show a surprisingly high prevalence of different anatomical findings on MRI in asymptomatic working-age adults (<60 years old), such as 40% prevalence of herniated discs (Weishaupt *et al.*, 1998), 54% prevalence of intervertebral disc bulging (Boden *et al.*, 1990), and 72% prevalence of degenerative intervertebral disc changes (Stadnik *et al.*, 1998). Thus, the fact that anatomical findings are present does not mean that the subject has symptoms. On the other hand, the fact that an injury is not found does not mean that it does not exist – the current diagnostic tools are not appropriate for identifying meaningful anatomical alterations and injuries (Wiker, 2003). We need new diagnostic tools or we need to adapt the existing ones so that we can assess potentially injured tissues such as ligaments and muscles or chemical changes.

Even for other types of WMSD such as wrist tendinitis, the diagnosis is based on clinical findings instead of histological evidence due to the invasiveness of the procedures that could actually "prove" the presence of injury (Palmer, 2003). Imaging is usually not useful and is often unnecessary for WLBD diagnosis (Deyo *et al.*, 1992). WLBD may occur due to biochemical alterations in the concentrations of proteoglycans, lactic acid, and/or pH alterations (Simmonds and Kumar, 1992). The presence of alterations is supported by strong evidence based on a review of systematic reviews of good quality of two or more studies (Van Tulder and Waddell, 2000) which looked at the use of non-steroidal anti-inflammatory drugs and muscle relaxants for alleviating WLBD in the acute phase. The effectiveness of anti-inflammatory drugs may be due to the anti-inflammatory and/or analgesic effects. In any case, it provides evidence of physical problems as opposed to psychosocial issues given that many of these studies are randomized control trials with a placebo group.

In 2005, overexertion was the most common cause for lost-time claims in all industries in Alberta; sprains, strains and tears were the leading nature of injuries, and the back was the most commonly injured body part accounting for over one-third of all claims (Alberta Human Resources and Employment – AHRE, 2006). Similarly, between 1991 and 2001, more than 40% of all injuries and illnesses in the U.S. (43.6% in 2001), resulting in days away from work, were sprains and strains; WLBD represented 24.2% of all lost time claims and 83% of the WLBD cases (N = 183,424) were classified as sprains and strains (N = 152,505) (BLS, 2001). The relationship between work biomechanical demands and WLBD has been established for many years by many studies (Bernard, 1997; Frymoyer *et al.*, 1980; Marras, 2000). There is sufficient evidence that awkward postures and heavy physical work and strong evidence that lifting/forceful movements and whole body vibration have a causal relationship with WLBD (Bernard, 1997). The following studies illustrate some of the evidence available in the literature. Chaffin and Park (1973) reported that high values for the ratio between the force exerted during work and the maximum isometric force of the worker are related to higher incidences of WLBD. Bending, twisting, lifting heavy weights and making forceful movements were shown to be related to WLBD (Frymoyer *et al.*, 1980).

Punnett *et al.* (1991) found in a case-control study (95 cases and 124 controls) that the risk of WLBD increased significantly due to prolonged bending and twisting during work (odds ratio – OR of 4.9 for mild trunk flexion, 5.7 for severe trunk flexion, and 5.9

for trunk twist or lateral bend). Myers *et al.* (1999) found in an other case-control study (200 cases and 400 controls) of WLBD in municipal workers that high job strain (OR = 2.12, 95% CI = 1.28-3.52), body mass index (OR= 1,54, 95% CI = 1.08-2.18) and twisting, extended reaching, and stooping (OR = 1.42, 95% CI = 0.97-2.08) were significant risk factor for WLBD. Norman *et al.* (1998) studied more than 10,000 automotive assembly workers. When the authors compared a sub-group of 104 cases (with WLBD) with 130 controls (without WLBD), the stronger biomechanical risk factors for WLBD included peak shear force on L4/L5 (OR = 2.3), peak trunk flexion (OR = 2.4), peak trunk velocity (OR = 1.9), and peak compression force on L4/L5 (OR = 1.9), usual force on the hands (OR = 1.9). In a three-year prospective study including 861 workers, the work biomechanical demands were directly assessed and it was found that trunk flexion (relative risk – RR of 1.72 for flexion ≥60° for more than 5% of working time/day), trunk rotation (RR of 1.57 for rotation ≥30° for more than 10% of working time/day), and weight lifting (RR of 1.79 for weight ≥25 kg for more than 15 times/day) were risk factors for WLBD (Hoogendoorn *et al.*, 2000).

Low back pain resulting in WLBD may be due to injuries to the tissues of the lumbar spine region such as ligaments, tendons, facet joints, vertebral periosteum, paravertebral muscles and fascia, blood vessels, intervertebral disc annulus fibrosus, or spinal nerve roots (Deyo and Weinstein, 2001). As illustrated in the previous paragraph, physical exertion in the workplace can result in the precipitation of WLBD. However, some investigators argue that WLBD is not caused by injury because, by definition, an injury precipitates at a specific time and place, and WLBD commonly have a gradual onset (Hall *et al.*, 1998).

Considering the Cumulative Load Theory of musculoskeletal injury precipitation, an injury may occur even without an accident due to an unusual activity, but due to performing a regular activity for a long period of time (Kumar, 2001). For example, the tendons stretch from 1% to 2% when load is applied (muscle contraction). After the contraction, the tendon usually returns to its initial length. However, if the contraction is repeated without enough recovery time or if the contraction is sustained for prolonged time, the tendons stay elongated by about 1% (residual strain) (Abrahams, 1967). This alters the mechanical efficiency of the muscles that may require an increase in contraction to generate the same amount of force. The tendon will have reduced stress tolerance capacity due to the residual strain (decreased cross-sectional area). Similar changes may be observed in other musculoskeletal tissues due to their viscoelastic characteristics. Thus, injury may occur without apparent changes in external workload or overexertion due to the cumulative effect of physical exertion.

There is some confusion in relation to the role of psychosocial issues in WLBD causation. The psychosocial issues play a more evident role in the level of disability and return to work after injury than in the injury precipitation itself (Feldman, 2004; Snook, 2004). Buchbinder *et al.* (2001) presented the results of a public campaign delivered by the media to alter beliefs about back pain. The campaign was intended to reduce disability and compensation costs in the state of Victoria in Australia showed that providing positive messages about back pain resulted in more positive beliefs about it, decreased the number of claims and days compensated, and reduced disability and compensation costs. However, the incidence of new cases was not positively affected. Actually, their data shows that the percentage of the sample that reported to have had "back pain during the last year" and "during the last week" increased respectively from 54% to 65.6% and from 27.3% to 36.8% between the first (August, 1997) and fourth (December, 2002) surveys (Buchbinder and Jolley, 2005). One should not assume that the factors influencing disability and return to work are the same factors causing injuries in the first place. "Injured workers commonly note that the attitude and manner of their supervisor or employer became far more negative after a reported injury" (Feldman, 2004).

None of the studies evaluating psychosocial risk factors have properly evaluated physical risk factors simultaneously (Davis and Heaney, 2000). It is hypothesized that psychosocial factors increase stress levels resulting in increased muscle contraction resulting in higher physical load (e.g. higher compression at L5/S1). Davis and Marras (2003) studied the contribution of biomechanical, psychosocial, and individual characteristics for spine loading. They found that the biomechanical demands of the task were the critical factors (load placement: 4% to 30% of explained variability, load weight: 15% to 55%, body dimensions – shear forces: 12% to 58% and compressive forces: 3%), and gender had some contribution (0.7% to 13.4%). However, psychosocial variables contributed much less to the spinal loads (personality: 6% to 19%, mental concentration and social environment: $\leq$0.2%). "Studies show that psychosocial factors influence back pain behaviour but are not important causes of pain itself" (Adams and Dolan, 2005).

1.4.2. Definition

The paradigm of this thesis involves primary prevention of WLBD including the mechanical model of degradation where WLBD precipitate through the steps of activity, biomechanical stress, temporal loading, tissue strain, and injury (Kumar, 2001; Marras, 2000; McGill, 1997). Thus, in this thesis, WLBD are defined as disorders whose precipitation is related to work demands and are connected to a physical problem causing dysfunction due to disruption of tissue, affecting the lumbar region of the spine, classified as ligament sprain, muscle strain, facet joint, and/or sacroiliac joint injury, resulting from over or cumulative exertion due to lifting, pushing, pulling, carrying, bending, and/or twisting during work.

1.4.3. Control

In this thesis the term WLBD control is used as opposed to prevention because of the fact that, given the multifactorial nature of WLBD precipitation, it is not feasible to completely prevent/eradicate WLBD. Despite, its control may be improved and a reduction may be achieved.

Efforts to improve work safety, efficiency, and productivity are long desired goals. Frederick Winslow Taylor is cited as being the pioneer of the "systematic study of work". The purpose of Taylor's first study in 1889 was to determine the ideal load-shovel size combination for shovellers (workers who shoveled pig iron at an industry called Bethlehem Steel). The result of Taylor's 3-year long study caused the productivity to increase close to 400% at the cost of many injuries to the workers and the entire crew had to be replaced every six months. After Taylor, during the first two decades of the 20th century, the Gilbreths (Frank and Lillian) conducted several motion and time studies. One of their works involved the assessment of their children movements during dish washing. They introduced the idea that there is a "best way of doing any job" (Kanigel, 1997). However, these efforts were focused on improving productivity and efficiency of production with no consideration for the worker safety.

Since the industrial revolution, many risks to the development and aggravation of WLBD have been identified and controlled to some extent. A common strategy used by industry has been the mechanization of production, but many activities inevitably require handling (Marras, 2000). Automation may help to alleviate the problem, but it causes another social problem, that of unemployment. Another issue is that in semi-automated systems, the task becomes more repetitive in nature and the workers feel "robotized" due to the fragmentation of the task (Coury *et al.*, 2000; Fredriksson *et al.*, 2001).

To improve the control of WLBD, the complexity of the interaction between man and his work has to be understood. By doing so, it might be possible to recognize the problematic aspects that allow interventions without compromising other relevant issues. WLBD are multifactorial in nature (Kumar, 2001). Musculoskeletal safety can be improved by controlling for risk factors, currently it is most feasible to control for the biomechanical, and to a lesser extent for the psychosocial, risk factors.

To determine the position that maximizes the biomechanical advantage of muscles and the safe ranges in which work can be accomplished with relatively low risk, it is necessary to quantify and classify the work biomechanical demands. Information about work biomechanical demands needs to be collected and analyzed in a systematic way in order to contribute to a deeper understanding of the relationship between physical exertions at work and WLBD. This procedure may result in a better control of WLBD by facilitating safe exertions through adjustment of the work biomechanical demands. This information will help to improve the prevention as well as the rehabilitation of these highly prevalent disorders. It is important to implement WLBD control programs, including orientation, training, addition of assistive devices and design of workstations and tools, in worksites having jobs involving stressful postures, forceful movements, and heavy manual materials handling. These exertions are often seen at hospitals and manufacturing companies.

1.5. Purpose of this PhD Study

A successful ergonomic intervention may result in decreased work-related musculoskeletal disorders (Occupational Safety and Health Administration – OSHA 1999). However, to achieve this major goal, the work biomechanical demands and their consequences need to be better understood. Therefore, the measurement, classification, and evaluation of the body's physical exposure to posture, movement, repetition, force, and the duration of the exertion are important to understand and control the work-related musculoskeletal disorders. The study of these factors is meant to assist in establishing ergonomic guidelines for safe work performance. It can contribute to better musculoskeletal health at work by reducing biomechanical hazards. The presented reasons show the need, justification, and call for further research in work biomechanical demands and work-related musculoskeletal disorders in order to improve their control.

The most common body part affected by work-related musculoskeletal disorders is the low back. Work-related low back disorders have many causes, incur large costs, and cause a great deal of suffering. A possible cause of this problem may be that the biomechanical demands of the work are too high, exceeding the normal capacity of the worker. For these reasons, the overall purpose of this research was to develop, test, and present a methodology that could be used in different worksites for control of work-related low back disorders in jobs and their risk factors. The major questions were "which are the risk factors, what are their magnitudes, and what are the possible improvements to improve control over the precipitation of the work-related low back disorders?" Three worksites (two steel companies, and a hospital) were selected based on having heavy jobs involving manual materials handling, and based on the higher incidence rates reported in the literature for these sectors of activity. The specific objectives of this research are presented as bullet points on the next page.

The specific objectives were to:

- Identify the jobs with higher incidence of work-related low back disorders in two steel companies and one hospital;
- Evaluate the problems related to work-related low back disorders and possible improvements for the highest incidence rate jobs based on the workers opinion;
- Identify the critical tasks of the jobs selected;
- Quantify the biomechanical demands (forces, repetitions, postures, movements, and duration) of specific (critical) job tasks;
- Evaluate possible associations between the biomechanical demands of the jobs and work-related low back disorders rates; and
- Propose/suggest evidence based adjustments, modifications, and/or training programs to reduce the biomechanical demands, to improve safety, and to reduce the risk of work-related low back disorders in these jobs.

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Chapter 2 – Working Postures

The initial version of this chapter was written during the directed individual research course I took with Dr. Kumar. In June 2004, a version of this chapter was published in the Journal of Occupational Rehabilitation (Vieira and Kumar, 2004). The chapter reviews some of the devices used to measure working postures and movements and focuses of the relationship between working postures and musculoskeletal disorders in general. The effects of working postures and movements are evaluated for specific body parts including the neck, upper extremities, the back, and the lower extremities. This chapter gives an overview of the relationship between postures/movements and work-related musculoskeletal disorders (WMSD), while the focus of the thesis is specifically on work-related low back disorders (WLBD).

2.1. Introduction

The improvement of musculoskeletal health at work is one of the most important objectives of ergonomics (Westgaard and Winkel, 1997). According to the International Ergonomics Association (IEA, 2000), physical ergonomics issues include "working postures, materials handling, repetitive movements, work-related musculoskeletal disorders, workplace layout, safety and health". Working postures and movements are important variables to be taken into account for occupational safety because they are two of the important mechanical variables and load determinants. Posture is defined in various ways considering the biomechanical alignment, the spatial arrangement of body parts, the relative position between segments, and the body attitude assumed to perform tasks (Haslegrave, 1994; Magee, 2002; Rohmert and Mainzer, 1996). Posture is influenced by the task, workstation, working tools' design, and the anthropometric characteristics of the workers (Coury, 1999).

Traditionally, working postures and movements have been widely assessed subjectively using various observational and graphical protocols and checklists such as the Ovaco Working Postures Assessment System – OWAS (Karhu *et al.*, 1977), the Rapid Upper Limb Assessment – RULA (McAtamney and Corlett, 1993), and the Rapid Entire Body Assessment – REBA (Hignett and McAtamney, 2000). All of these assessment tools use on-the-job observation or video recordings to categorize the ranges within which each body segment falls. The classification is based on raters' subjective judgments of joint angles. These protocols have evident limitations in the characterization of physical exposure. Among the problems are issues of subjectivity, rater bias, low precision, long periods of analysis, and the requirement for highly trained observers (Yen and Radwin, 2002). Their internal and external validity have also been questioned (Juul-Kristensen *et al.*, 1997).

A study conducted by Brodie and Wells (1997) showed that the results of the RULA, the Occupational Safety and Health Administration (OSHA) checklist for analyzing stress on the upper limb and lower back (OSHA, 1999), and the neck, trunk, and legs checklist (Keyserling *et al.*, 1992), and the upper extremity checklist (Keyserling *et al.*, 1993) are very unreliable and inaccurate. On the other hand, studies that use quantitative biomechanical measures are more precise and reliable. An advantage of direct measurements is that they provide detailed and accurate values for jobs with different work tasks (Juul-Kristensen *et al.*, 2001). But the measurement devices are more complex. They may influence performance of the task, increasing the chances of the activity not been typical work (Coury, 1999). Also, the amount of data generated by direct measures is extensive, and managing all this information is difficult. Despite, Yen and Radwin (2002) found that a graphic protocol required 6.3 times more analysis time than measures taken with electrogoniometers (elgons). Even when the time required for device attachment and sensor calibration was taken into account, the total time for data collection and analysis using elgons was 23% less than that for the observational method. Also, the difference in mean joint angle was 10.3 times smaller using elgons than for the observational method: 0.8° (0.59) vs. 11.4° (1.58), respectively. The root mean square (RMS) joint angle difference among analysts was 0.9° (0.61) using elgons and 7.1° (2.53) for the observational method.

Working postures and movements are addressed in almost every paper in the ergonomics field, but surprisingly, there seems to be only one book that specifically concentrates on these aspects of human kinematics (Corlett *et al.*, 1986). Furthermore, this book only links the research reports presented at the First International Occupational Ergonomics Symposium (Zadar, Yugoslavia, April 15–17, 1985). Also, scientific papers dealing with working postures and movements themselves are not common; knowledge has been elusive. For these reasons, the objectives of this paper were (I) to review the working posture literature, gathering information about the methods most often used to assess working postures by direct biomechanical measures; (II) to condense the available information about the relationship between working postures and musculoskeletal health; and (III) to clarify how working posture surveys help in the improvement of the control of WMSD.

2.2. Methods

This review included articles that met the inclusion criteria stated below. Papers published in the English language before March 2003 including the phrase "working postures" in the title, abstract, or keywords were searched in the PubMed (Medline), Scirus (Elsevier Science), and Science Direct databases. To be included, the articles had to use direct quantitative measures of working postures, to have external validity (on-the-job studies or job-simulated studies) and internal validity (the use of validated methods), and to include seven or more subjects. The choice of seven subjects as the cut-point was based on the fact that several of the papers reviewed and considered as relevant included at least that number of subjects. Relevant papers in the reference lists of those articles selected as well as review papers connecting the results of articles meeting the established criteria and pieces of information from relevant textbooks were also included.

2.3. Results and Considerations

2.3.1. Recently Published Articles on Working Postures

Selected articles published between 2000 and 2002 on working postures that used quantitative biomechanical measures to assess directly physical stress exposure and that focused on the assessment of working postures were reviewed. The studies were generically classified in those that dealt with static postures, quasi-static posture, quasi-dynamic postures, and dynamic postures and are presented in Tables 2-1 to2-4 (one table for each category). The references, occupations, posture descriptions, measuring methods, posture-related results, recommendations, and limitations or weaknesses of the studies were presented.

Table 2-1. Studies of static postures

Reference	Occupation Assessed	Posture Measuring Method	Posture Description	Posture-Related Results	Recommendations	Limitations or Weaknesses
Delleman and Berndsen (2002)	Computer work	Opto-electronic videogrammetry (VICON system)	Seated performing manual data entry.	Backrest backward inclinations between 0° and 15° did not interfere with task performance.	The VDU[a] gaze inclination should be between 6° and 9° below the horizontal.	Too short postural exposure duration.
Szeto *et al.* (2002)	Computer work	2-D motion analysis (Peak 5 system)	Seated performing manual data entry.	Symptomatic workers tend to have a more forward head posture than asymptomatic ones.	Not included.	The results did not reach statistical significance.
Moffet *et al.* (2002)	Computer work (laptop)	Videogrammetry, elgons[b], EMG[c]	Seated performing manual data entry. Lap vs. desk situations	Desk vs. lap work: < neck-trunk flexion, wrist extension, but > shoulder elevation.	The work situation (lap vs. desk) should take into account when making recommendations.	Sample of eight did not allow identifying differences between laptop designs.
Bonney and Corlett (2002)	Computer work	Precision stadiometery and inclinometry	Seated looking at a screen in a controlled static position.	Cervical spine shrinkage was bigger with neck flexion (20° and 40°) than in neutral position (0°).	Neural neck posture was recommended for long VDU work.	Subjects were watching a movie on the VDU. This may not represent common VDU work.
Andreoni *et al.* (2002)	Car driving	3-D optoelectronic system (ELITE) and matrices of pressure sensors (TEKSCAN)	Seated with the hands on the steering wheel.	The method allows for the measurement of car driver posture parameters and identifies specific sitting strategies.	The method should be used in mock-ups / commercial cars to obtain kinetic and kinematic data for guidelines.	The study addressed only static analysis.

[a] VDU – visual display unit.
[b] elgons – electrogoniometers.
[c] EMG- electromyography.

Table 2-2. Studies of quasi-static postures

Reference	Occupation Assessed	Posture Measuring Method	Posture Description	Posture-Related Results	Recommendations	Limitations or Weaknesses
Juul-Kristensen *et al.* (2001)	Poultry processing work	Videometry, inclinometers, EMG[a] and goniometers.	Standing with neck flexion, working with the upper limbs, performing manual deboning or packing chicken.	% of time spent in neck flexion >20° differed between methods. Wrist posture was similar: 0° of flexion and 19° of ulnar deviation.	Observation-based methods should use more general classifications, especially for hand positions.	Possible interference in the task performance due to the many devices attached to the subjects.
Fredriksson *et al.* (2001)	Industrial car sealing work	Videometry (PEOFlex), Physiometer	Forward-bent position with the head and trunk doing repetitive work with the upper limbs.	Decrease in neck and trunk flexion and repetitive work with hands, but work-cycle time decreased and WMSD[b] increased.	Interventions should account for work-cycle time and decreases in occupational pride/ psychosocial issues.	Shorter work-cycles may have caused increased WMSD, not psychosocial issues.
Stal *et al.* (2000)	Machine milking work	EMG, strain gauge force transducer	Milking cows using two different systems with shoulders flexed and elbows extended.	Traditional tethering system: > peak loads. Modern loose-housing system: > productivity, but >static load and <rest.	Newly adapted techniques for female milkers are necessary to reduce the risk of upper extremities WMSD.	Dynamic EMG may contain movement artifacts.
Palmerud *et al.* (2000)	Industrial manual handling work	3-D Videogrammetry (Qualisys system), and intramuscular pressure	Manual lifting with different arm postures and hand loads.	Bigger intramuscular pressure as the arm was elevated from the neutral position (0°) to 90° of both flexion and abduction.	Particular attention should be placed on activities that require handling of heavy hand powered tools in elevated positions of the arms.	The simulation task did not have a specific aim to be achieved as in industrial work.

[a] EMG – electromyography.
[b] WMSD – work-related musculoskeletal disorders.

Table 2-3. Studies of quasi-dynamic postures

Reference	Occupation Assessed	Posture Measuring Method	Posture Description	Posture-Related Results	Recommendations	Limitations or Weaknesses
Gallagher *et al.* (2002)	Mining work	Elgons[a] (LMM[b]), EMG[c], Force Plate.	Kneeling, stooping and standing while lifting an electrical cable 7.6 m long.	Variation in posture influenced muscle recruitment and spinal load.	Kneeling on both knees was the least stressful posture, but mechanical assistance is recommended.	The laboratory simulation did not account for the space constraints of real work.
Ferguson *et al.* (2002)	Storage work	Elgons (LMM), EMG, Force Plate.	Lifting a 11.3 kg box from an industrial storage bin standing on 1 or 2 feet, using 1 or 2 hands, and with or without supporting body weight.	Using 1 hand supported lifting, standing on both feet, minimized spinal load the most.	A bin design facilitating lifting from the upper front with handholds on the side.	The subjects were students with no manual material handling experience.
Bjoring and Hagg (2000)	Painters' work	Fluid-based angle transducers	Painting pieces on a horizontal surface abducting the right upper-arm and griping a spray gun.	The upper-arm abduction when painting was high, as well as the gripping force.	Installation of tables with height control and changes in the spray gun.	Effect of the recommendations was not evaluated.
Rose *et al.* (2001)	Constructio n work	A handle connected to a strain gauge and BM (JACK)	Standing working on floor level with fully flexed trunk.	The mean endurance time was 6.5 min for the unskilled and 10.75 min for the rodmen.	Fully flexed postures may be assessed by general prediction endurance models.	The joint angles were not measured.
Hsu and Chen (2000)	Filing work	Videogrammetry (Qualisys system).	Standing working with the upper limbs filing a plate with trunk and neck semi-flexed.	The right lower arm and hand are maintained in a more neutral position with the new file.	The bent-handled file (60°) should be used instead of the flat-handled file.	The postural measures were taken only for the right arm.

[a] Elgons – electrogoniometers.
[b] LMM – lumbar motion monitor.
[c] EMG – electromyography.
[d] BM – biomechanical model.

Table 2-4. Studies of dynamic postures

Reference	Occupation Assessed	Posture Measuring Method	Posture Description	Posture-Related Results	Recommendations	Limitations or Weaknesses
Caboor *et al.* (2000)	Nursing	Elgons[a], EMG[b], inclinometry	Standing and handling patients in beds.	EMG and range of motion was not influenced, but the time spent more erect increased.	Bed height should be adjustable.	"Voluntary" adjustments of bed heights
Mirka *et al.* (2002)	Furniture manufacturing (upholsterers and machine room work).	Elgons (LMM[c])	Working with wood using the upper limbs, torso, head and, eventually, lower limbs.	Upholsterers: flexion > 50°; lateral bending and twisting >20°. Machine room: flexion >80°; lateral bending > 15°; twisting >45°.	The height-adjustable upholstery buck and the machine room lift systems should be used as engineering controls.	The subjects were students not used to the tasks and not under time pressure.
Schibye *et al.* (2001)	Refuse collectors' work	3-D videogrammetry (Peak Motus), BM (Watbak) and force transducers.	Pushing and pulling waste containers.	Pushing/pulling compression at L4/L5: 605-1445N. Shear force < 202N. The torque at the shoulders: 1-38 Nm.	Pushing and pulling should be used instead of lifting and carrying whenever possible.	Used a static BM for a dynamic task.
Lavender *et al.* (2000)	Firefighters' work	Elgons (LMM), videogrammetry, BM[d] and hand-held dynamometer.	Transferring a patient from bed to a stretcher, carrying down stairs, around a landing, and transferring to a hospital gurney.	Victims were lifted with of the trunk and shoulder flexion.	The transferences should be done using an interface board. The leader should walk facing forward when descending stairs.	Simulated tasks: no time pressure, light dummy was carried and transferred.
Kuijer *et al.* (2000)	Refuse collectors' work	Videogrammetry (TRAC-system), VO2, heart rate (PE 4000).	Collecting refuse in bags and pushing and pulling two-wheeled containers.	Working postures and perceived exertion in 1997 were more favorable than in 1993.	Introduce job rotation and effective work–rest schedule as well as redesign the two-wheeled containers.	Different populations from different studies were compared.

[a] Elgons – electrogoniometers.
[b] EMG – electromyography.
[c] LMM – lumbar motion monitor.
[d] BM – biomechanical model.

2.3.2. Methods of Direct Assessment of Working Postures

Posture and movement measurements are important factors in the determination of normality and variability, as well as in the physical examination process (Norkin and White, 1995). In addition, these measurements are an essential part of the estimation of risk rates for the development of WMSD (Marras *et al.*, 1999). Posture and movement measurement properties need to be adequate including their reliability and validity (Gilliam *et al.*, 1994; Mayer *et al.*, 1997). Reliability and validity are attributes of measurements, not of devices, and depend on the conditions, purposes, and contexts in which the measurements are taken (Portney and Watkins, 2000). While reliability is related to stability, consistency, and lack of random error, validity deals with accuracy, correctness, and the absence of systematic error, as well as the ability to make inferences (Sim and Arnell, 1993). For a review of the measurement properties and terminology adopted in this thesis refer to Gadotti *et al.* (2006).

A brief review of the measurement devices most frequently used to directly assess working postures and movements is presented. It focuses mainly on the reliability and validity of the measurements taken with the different devices. They are presented by type, and the order of presentation is in accordance with the degree of complexity of the method and the time of its introduction. A description of the method or measurement device is presented along with results of specific studies about the reliability and validity of the measurements from the devices in specific conditions. For a broader review of the different working postures' recording methods, including graphic protocols, questionnaires, and checklists, the reviews conducted by Rohmert and Mainzer (1986), Pinzke (1996), Coury (1999), and Li and Buckle (1999) are recommended.

Goniometers: They are the simplest devices to measure posture and movement in degrees (Gajdosik and Bohannon, 1987), and they are the most used tool to quantify range of motion (Domholdt, 2000). According to Miller (1985), goniometers have been used since, at least, 1910. In 1949, they were termed "universal goniometers" because it was thought that they could be employed to quantify the movement of all joints. Several goniometers made of wood or metal were developed in France, where they were used to quantify the dysfunctions of combatants from the First World War. In 1952, the transparent goniometer was developed in order to improve the accuracy of axis alignment.

The second edition of the American Medical Association (AMA, 1987) guide to evaluation of impairment recommends the utilization of long arm goniometers to measure spine motion in the sagittal and frontal planes. However, this recommendation was changed in the fourth edition of the guide (AMA, 1993), which recommends inclinometers as the appropriate tool. Gilliam *et al.* (1994) tested the reliability and validity of goniometric measurements of pelvic angle. The authors found high intra- and inter-tester reliability (intra-class correlation coefficients – ICCs from 0.93 to 0.96 for intra and ICC = 0.95 for inter-tester correlation. With respect to concurrent validity, a reasonable correlation was found between radiographic and goniometric measurements of pelvic angle (Pearson correlation coefficient – r from 0.68 to 0.85).

Universal goniometers have been criticized in relation to their validity because they have a single-hinge joint, while the axes of human joints change position during movement (Miller, 1985). Thus, the biomechanical complexity of spine motion necessitates caution when using goniometers to measure spine movement since the data can be affected by errors. Universal goniometers are not the ideal tool to measure movements made by multiple joints with dynamic axes (Tesio *et al.*, 1995). Domholdt (2000) presented the possible sources of error associated with goniometric measurements. The main categories are instrument problems, rater errors, inconsistency among raters, and patients'/subjects' fluctuation in performance. Nitschke *et al.* (1999) tested the inter- and intra-tester reliability of goniometric and inclinometric measurements of lumbar movement. A mean error of 8° was found among tests, and the mean error found among testers was 14.16°. The authors argued that impairment ratings based on subjects' range of motion could vary from 0% to 18% due to the unreliability of the measurements of both devices.

Inclinometers: They are, and function as pendulum goniometers or gravity dependent goniometers (Norkin and White, 1995; Williams *et al.*, 1993). When one inclinometer is used to measure spine movements, the range recorded represents the sum of the spine, pelvic, and hip movements, and thus the two-inclinometer method should be preferred.

The two-inclinometer method isolates spine movement by subtracting the value registered by the lower inclinometer from the value registered by the upper one (Williams *et al.*, 1993).

Leighton first introduced inclinometers to measure joint motion in 1955. He conducted a study comparing inclinometer movement measurements of the spine and upper and lower extremities with those of universal goniometers and found good parallel reliability (r from 0.913 to 0.996) (Leighton, 1955). In contrast, Gill *et al.* (1988) found low instrument reliability with the two-inclinometer method (lower inclinometer: coefficient of variation – CV = 33.9%, upper inclinometer: CV = 9.3%). Similarly, Rondinelli *et al.* (1992) found that the measurements taken with inclinometers have low intra-, inter-tester, and parallel reliability.

Miller *et al.* (1992) stated that inclinometry is not an ideal method to assess spine motion. Mayer *et al.* (1995) found low inter-tester reliability for inclinometric measurements of lumbar motion (mean error of 8°). The intra-tester and inter-tester reliability of the two-inclinometer method was also assessed by Williams *et al.* (1993), they found low reliability (intra-tester reliability: r from 0.13 to 0.87 for flexion and from 0.28 to 0.66 for extension, and inter- tester reliability: ICC = 0.60 for flexion and 0.48 for extension).

Photogrammetry: According to Miller (1985), the use of photographs as a way of quantifying joint position (photogrammetry) was proposed in 1964 by Wilson and Stasch. Photographs are important in both clinical and research environments because it is simple, noninvasive, and less expensive than radiographs (Chen and Lee, 1997; Corry, 1999). Photographs are used for recording body posture by attaching markers to specific bony landmarks to represent joints or body segments (Gajdosik *et al.*, 1994; Kumar, 1974).

Chen and Lee (1997) studied a photogrammetric technique to measure the lumbar posture; the authors found good concurrent validity (determination coefficient – r^2 from 0.91 to 0.98) by comparing its measurements with X-ray measurements of lumbar sagittal movement (Chen and Lee, 1997). The authors found differences between vertebral (measured with X-rays) and markers' orientations only to levels L5 and S1. In contrast, Gill *et al.* (1988) did not obtain acceptable results with another photogrammetric technique (CV of 6%). Thus, the validity and reliability of photogrammetric techniques depend on the specific procedures, methods, and type of skin marker used.

Potentiometric Electrogoniometers (elgons): They are able to advance the study of motion by recording of continuous motion (Nicol, 1987). Potentiometric elgons can have one or multiple axes. The one-axis elgon consists of two arms joined at the axis of a potentiometer that generates electrical signals proportional to the motion. As a function of its single axis, it faces the same problems as the standard goniometer (Tesio *et al.*, 1995). This type of elgon has been used since 1959; the angular displacement causes variation in the potentiometer resistance, resulting in voltage oscillations recorded as angular motion (Norkin and White, 1995; Punnett *et al.*, 1991).

There are a variety of multiple-axis elgons, and usually they record data in two or three planes simultaneously. Paquet *et al.* (1991) tested the concurrent validity of a home-made multiple-axis potentiometric elgon to measure spine movement. The measurements of a previously validated inclinometer were used as a reference. The instrument's reliability (test–retest) was also assessed. Good results were found for both aspects (r = 0.97, 1.03 slope and ICC = 0.982, respectively). Perrtet *et al.* (2001) tested the concurrent validity of another potentiometric elgon (the Rachimetre) using radiographs as a criterion measure of pelvic mobility. The authors found a Spearman correlation coefficient of 0.89, and an ICC of 0.65. On the basis of these results, they concluded that the validity of the Rachimetre measures was acceptable. Lee *et al.* (2002) assessed lumbar spine movement in the sagittal plane with an in-house-developed three-axis potentiometric elgon and compared the measurements with those registered by video-fluoroscopy (a cine-radiographic imaging technique). The elgon measurements presented high concurrent validity (mean error = 0.61°, SD = 0.28°).

The Orthopaedic Systems Inc. elgon (Union City – CA, US), named the OSI CA-6000 Spine Motion Analyzer, also uses potentiometers to record movement. It has been used to measure both thoraco-lumbar (Feipel *et al.*, 2001) and cervical movement (Christensen and Nilsson, 1998; Feipel *et al.*, 1999; Lantz *et al.*, 1999). Christensen (1999) evaluated the instrument's reliability and the concurrent validity of the CA-6000 elgon in relation to two protractors. The measurements results were within 0.01° of error, showing high instrument reliability. However, the disagreement between the elgon and the protectors was from 2% to 11.5%, showing low parallel reliability. Thoumie *et al.* (1998) studied the concurrent validity of measurements of the lumbar sagittal motion of 12 healthy subjects; they found that the correlation between X-ray and potentiometric elgon measurements was moderate (r from 0.58 to 0.77).

The lumbar motion monitor (LMM) permits recording velocity and acceleration along with motion of the torso. It is another potentiometric elgon that has been used to analyze lumbar spine movement (Marras *et al.*, 1992). It was used to test whether motion characteristics could be used to assess low back pain in a sample of 709 subjects. Patients with low back disorders were correctly distinguished from healthy subjects in 88% to 90% of the cases (Marras *et al.*, 1999), showing that spinal kinematic measurements have high predictive validity for classifying low back disorders. However, the LMM is an exoskeleton that may interfere with the tasks being performed and it is not ideal for workplace setting studies. Lavender *et al.* (1999) compared five methods for determining the risk of work-related low back disorders (WLBD) and found that the LMM is likely to overestimate the risks involved in jobs. Thus, this method is more appropriate to evaluate patients than to predict risk of WLBD.

Flexible Electrogoniometers (elgons): Flexible elgons have also been used to record posture and movement continuously (e.g. Boocock *et al.*, 1994). According to Tesio *et al.* (1995), flexible elgons avoid biomechanical problems related to axis alignment since they have no axis, improving the validity of the measurements. Thus, flexible elgons are preferred to potentiometric elgons. The Biometrics Ltd. (Gent, UK) manufactures flexible elgons that have two endblocks linked by a wire with strain gauges placed in four orthogonal planes. The wire is covered by a spring that slides inside the endblocks, allowing linear displacement during movement. The recorded angle refers to the orientation between the endblocks (Biometrics, 1999). Boocock *et al.* (1994) tested the reliability and practicality (usability and acceptability) of a flexible elgon. The authors compared measurements of lumbar movement in the sagittal plane using a flexible elgon with those of an inclinometer and a flexicurve and found good parallel reliability. The authors also used the flexible elgon to quantify the lumbar movement of four garage mechanics during a 2h period of regular work and concluded that the device is useful to investigate spinal kinematics in workplace settings.

Video Analysis Systems: Video analysis systems record or automatically track surface markers placed over the skin. The subject is filmed during activity and the Cartesian coordinates of the markers are tracked or determined through digitization of the points (videogrammetry) (e.g. Brumagne *et al.*, 1999).

Other methods, such as opto-eletric (Domholdt, 2000) and magnetic tracking systems (Robb, 1999), utilize light-emitting diodes or magnetic trajectory systems, respectively, dispensing with digitization procedures. Video analysis systems allow for the assessment of the angles between line segments specified by the markers and make possible 3-D assessment of body kinematics. The results of a study by Brumagne *et al.* (1999) showed high parallel reliability for a 3-D video analysis system. Measurements of lumbosacral movement were compared with those from an elgon (r - 0.84 to 0.97). Gracovetsky *et al.* (1995) found good concurrent validity for videogrammetric measurements of healthy subjects' lumbar motion in the sagittal and frontal planes when compared to radiographic measurements.

2.3.3. The Relationship between Working Postures and Musculoskeletal Health

Loading the musculoskeletal system can have both immediate and cumulative effects and may account for dysfunctions later in life depending on the magnitude and frequency of loading (Kumar, 1990). Awkward, constrained, asymmetric, repeated, and prolonged postures can overload tissues and exceed their thresholds of tolerable stress, causing injury due to overexertion or imbalance (Kumar, 1994). The maintenance of considerable static postures for prolonged periods of time compresses the veins and capillaries inside the muscles, causing micro-injuries associated with prolonged reduction of tissue oxygenation and nutrition. All these factors can cause imbalance, fatigue, discomfort, and pain due to disruption of tissues (Kumar, 1990 and 1994).

Tissues that may be injured due to working postures are muscles, tendons, and ligaments. Nerves can be injured secondarily due to compression or ischemia. In addition, joints, bones and their cartilages can also be damaged by significant loads and strains accumulated over years (Kumar, 2001). A review conducted by Magnusson and Pope (1998) verified that posture is associated with problems in the neck, shoulder, arms, hips, and knees. In addition, NIOSH reviewed over 700 epidemiological studies and concluded that there is sufficient evidence that posture is related to back, shoulder, and wrist WMSD (Bernard, 1997). Therefore, there is relationship between working postures and WMSD. In the following sections, evidence of this relationship is presented for specific body parts.

Head and Neck: A forward posture of the head can be a resultant adaptation due to work exposure to tasks requiring this posture (Darnell, 1983). A two-year follow-up study conducted by Kilbom *et al.* (1986) showed a positive relationship between neck flexion and musculoskeletal symptoms at the neck ($r = 0.41$, $p < 0.001$) and at the trapezius muscle ($r = 0.28$, $p < 0.01$). Also, forward posture of the head was shown to be correlated to temporo-mandibular, neck and back pain (Rocabado and Iglarsh, 1991). Finsen *et al.* (1998) studied 93 dentists and found that between 50% and 60% of the neck pain cases were associated with prolonged neck flexion. The authors found that the dentists maintained more then 15° of neck flexion during 97% of the time, and maintained more then 30° of neck flexion during 82% of the time working with patients (OR = 7.0 for time between 25 and 30h work/week). Neck flexion was found to be more accentuated in symptomatic subjects than in asymptomatic ones (Szeto *et al.*, 2002).

Trunk: Back disorders have the highest incidence rate and represent the most costly WMSD (e.g., Department of Education, 1993; Nordin *et al.*, 1997). In an epidemiological study conducted by Kelsey and Hardy (1975), prolonged sitting in association with vibration was found to be a risk factor for low back pain. Bending and twisting were also shown to be related to low back pain (Frymoyer *et al.*, 1980). Intervertebral disk compression is at a minimum when laying down and at a maximum when sitting with the trunk bent forward. The intermediate compression positions are respectively: standing upright, sitting upright, and standing with the trunk bent forward (Nachemson, 1981).

Compression and shear over working life was shown to be higher in subjects with back pain (Kumar, 1990). Lifting heavy weights and making forceful movements are also related to low back pain (Punnett *et al.*, 1991). A review of the biomechanics and epidemiology of working postures presented evidence that both sitting and standing for long periods can trigger low back pain (Magnusson and Pope, 1998). In a 3-year prospective study including 861 workers, physical load was directly assessed, and it was seen that spinal kinematics was critical to the development and aggravation of WLBD. Trunk flexion (relative risk – RR of 1.72 for flexion ≥60° for more than 5% of working time/day), trunk rotation (RR of 1.57 for rotation ≥30° for more than 10% of working time/day), and weight lifting (RR of 1.79 for weight ≥25 kg for more than 15% of working time/day) were found to be risk factors for WLBD (Hoogendoorn *et al.*, 2000).

The cumulative effects of increased axial load and the force of gravity can cause the cervical lordosis and thoracic kyphosis to increase and the lumbar lordosis and trunk height to decrease (Figure 2-1).

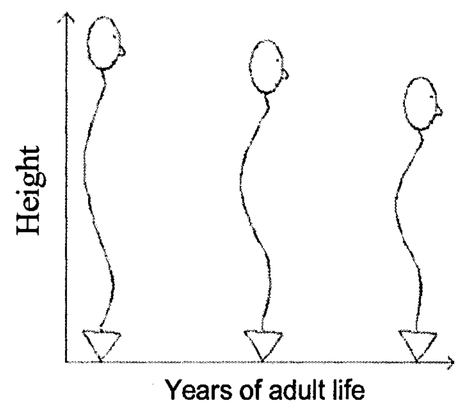

Figure 2-1. Postural adaptations due to cumulative axial load (reproduced from Vieira and Kumar, 2004).

The process illustrated is intensified by intervertebral disk dehydration and bone morphological rearrangements (Magnusson and Pope, 1998). These cumulative adaptations are determined by a combination of mechanical exposure, personal habits, lifestyle, and psychosocial factors. Some of these postural modifications are temporal adjustments, but others are permanent. These lasting and cumulative alterations are often found in elderly people (Hammerberg and Wood, 2003). Cumulative postural and morphological changes in the spine result in altered biomechanical loading of back muscles, ligaments and joints, and can precipitate back disorders. Nahit *et al.* (2001) studied 1081workers from 12 occupational groups and found that posture was not as critical as weight, manual handling, and repetitive movements were to the precipitation of pain in newly employed workers. Thus, the hazards of inadequate working postures and overloading are more likely to appear among those who are exposed to harmful postures for longer periods of time.

Upper Limbs: A study by Bjelle *et al.* (1979) showed that close to 70% of industrial workers that presented shoulder pain worked with their hands at or above shoulder level. Wiker *et al.* (1989) found that working with hands above shoulder level significantly increased the risk of localized muscle fatigue (increased postural tremor) and reported postural discomfort even when the external loads on the hands were light and the subjects young and robust. Finsen *et al.* (1998) found that the 88% prevalence of shoulder pain in dentists was associated with shoulder abduction of more than 30° for 1/3 of the time working with patients.

Ulnar deviation, wrist extension of more than 45°, forceful pinch and grip, and high repetitiveness (work-cycles with durations of less than 30 s and repetitions of the same movements for more than 50% of the working day) were found to significantly increase the risk of hand and wrist WMSD (Silverstein *et al.*, 1986a and b). Palmerud *et al.* (2000) studied the intramuscular pressure of the infra- and supraspinatus muscles in different arm postures and with different hand loads. The authors found a linear increase in the intramuscular pressure as the arm was elevated from the neutral position (0°) to 90° of both flexion and abduction. For the supraspinatus, the increase was mainly determined by the posture and not by hand loads. Buckle and Devereux (2002) found strong evidence that the combination of posture, repetition, force, and vibration is a risk factor for upper limb WMSD.

Lower Limbs: WMSD affecting the lower limbs are less common than those involving the back, neck, and upper limbs (Li and Buckle, 1999). Nevertheless, in sitting, the pressure against the hamstrings and gluteus is increased in the absence of feet support. It can cause compression, ischemia, and accumulation of metabolites (Magnusson and Pope, 1998). Nahit *et al.* (2001) found that workers performing work in kneeling posture for 15 min or longer had higher incidence of knee pain (OR = 1.8).

2.4 Discussion and Conclusions of Chapter 2

There is no ideal posture; even low levels of continuous muscle contraction represent a risk for causing WMSD is sustained for prolonged periods of time (Westgaard *et al.*, 1986). Long-lasting static postures are not recommended because they will result in discomfort (Li and Buckle, 1999). Thus, safe levels of "skeletal-muscle monotony" (continuous isometric contractions or low tension and high frequency repetitiveness) must be established (Sjogaard, 1986).

Changes in posture are present even during sleeping; an ideal workstation would allow for postural changes in order to permit working within comfortable and safe conditions (Andersson, 1986). Posture influences the strength that muscles are able to generate (Kumar *et al.*, 1991). The layout of workstations, tasks, and tools used also influence the amount of physical load that workers are exposed to. Acting to control injuries prior to their precipitation can help to avoid the onset of disorders and dysfunctions.

To conduct a successful ergonomic intervention, it is necessary to optimize musculoskeletal functioning and safety, while keeping production and its costs within manageable margins. de Looze *et al.* (2001) summarized seven cases where interventions successfully achieved their goal of reducing the physical workload of scaffolders, bricklayers, bricklayers' assistants, roof-workers, aircraft loaders, glaziers, and assembly line workers.

By classifying and quantifying the details of working postures, one might be able to determine the position that maximizes the biomechanical advantage of muscles and the safe ranges in which work can be done with relatively low risk. Thus, it is possible to better control for WMSD by facilitating safe postures. This procedure involves postural orientation, workers' training, and design of workstations and tools. Information about working postures needs to be collected and analyzed in a more systematic way in order to contribute to a deeper understanding of the relationship between working postures and

WMSD. This information will help to improve the control and rehabilitation of these high prevalence disorders.

On the basis of the published literature, it is clear that inappropriate working postures produce harmful physical exposures that can cause musculoskeletal injury, pain, and kinematic disorders. The multivariate interaction, overexertion, differential fatigue, and cumulative load theories explain the precipitation of musculoskeletal injuries (Kumar, 2001). The non-biomechanical factors affecting the precipitation of WMSD are anthropometric, genetic, and psychosocial predispositions.

The study of working postures assists in the establishment of ergonomic guidelines for safe work, contributing to better musculoskeletal health at work by reducing biomechanical hazards and improving the control of the WMSD. In addition, deep understanding of the working postures may help to improve the rehabilitation program of injured workers. This information can help to design treatment programs specific to the demands of the workers' jobs. Thus, areas for future research include studying means to use assessments of the working postures of injured workers to improve their rehabilitation.

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Chapter 3 – Occupational Biomechanical Demand Evaluation

Based on the working postures review presented in the previous chapter, the need for a classification system for postures and movements as well as for the forces, repetition and duration of the work was identified. This chapter presents the classification system proposed. A version of this chapter has been published in the Proceedings of the 2nd Annual Regional National Occupational Research Agenda of the National Institute for Occupational Safety and Health (NIOSH) (Vieira and Kumar, 2004a).

3.1. Introduction

The maintenance of static postures and force exertion for prolonged periods of time compresses the veins and capillaries inside the muscles, causing micro-necrosis due to the absence of tissue oxygenation and nutrition (Kumar, 1990 and 1994a). These factors can cause fatigue, discomfort, and disruption of tissues activating the nocioceptors, and, as a result, the worker experiences pain and may begin to exhibit pain behavior such as kinesiophobia (fear of moving), resulting in prolonged dysfunction. The tissues that are injured due to overexertion are the muscles, their tendons, and the ligaments. Nerves can be injured secondarily due to compression or ischemia. In addition, the joints' bones and their cartilages can also be damaged by the load and strain accumulated over years of activity (the Cumulative Load Theory – Kumar 2001; the cumulative nature of disorders – Armstrong *et al.*, 1993; Buckle and Devereux, 2002).

The methods currently in use to evaluate work biomechanical demands such as the Ovaco Working Postures Assessment System - OWAS (Karhu *et al.*, 1977), the Rapid Upper Limb Assessment - RULA (McAttamney and Corlett, 1993), and the Rapid Entire Body Assessment - REBA (Hignett and McAtamney, 2000) use observation on-the-job or video recordings to categorize the ranges within which each body segment falls and the estimated loads handled. Also, direct measures such as goniometers, inclinometers, photographic techniques, electrogoniometers, and video recording systems are used to measure working postures, movements, and repetition. In addition, dynamometers are used to measure the force exerted during a job task, and electromyography is used to record the electric activity of some of the muscles contracting during work exertion

(Vieira and Kumar, 2004b). Independently of the recording method used, a classification system of the risk associated with work biomechanical demands is necessary.

By classifying and quantifying the details of the different physical exertions at work one might be able to know the position that maximizes biomechanical advantage of muscles and the safe ranges that work can be done with reduced risk. A quantitative classification system of the physical exertion determinants (posture, movement, repetition, force, and duration) would allow a better control of WMSD. For this reason, the objective of this paper was to develop and propose a system to quantitatively evaluate the work biomechanical demands. For conciseness, the present paper presents the proposed system using the movement dimension of work physical exertion as an example.

3.2. Methods

The need of an evaluation system for the work biomechanical demands was identified during a literature review focusing mainly on studies measuring working postures and movements, and on studies concerning the measurement devices' reliability and validity (Vieira and Kumar, 2004b). Based on the information gathered in the review, the present system of evaluation was developed and an initial report was presented (Vieira and Kumar, 2003). The process followed in identifying the most appropriate evaluation system for the movement dimension was based on the joints' range of motion (ROM), joints positioning, and angular motion covered during work. For each preliminary model developed, trials were conducted so that data from simulated work activities were entered in the models and checked in relation to their appropriate classification. The content and usefulness of the system was further considered and elaborated resulting in modifications of the initial idea and the final model is presented in this paper.

3.3. Results

The movement dimension demand of the work physical exertion can be classified using the movement demand index (MDI). The MDI is expressed as angular motions in relation to the ROM of each joint, divided by the number of joints taken into account (EQ 1).

Calculation of the Movement Demand Index (MDI):

$$MDI = \frac{100}{n}\sum_{j=1}^{n}\frac{AM_j}{ROM_j} \qquad \text{(EQ 1)}$$

Where, j = Joints 1…n, n = accounted number of joints, AM = Angular Motion, ROM = Range of Motion. ROM is the maximum range that a joint can actively cover and AM is the range that a joint covers during an activity.

This calculation results in a mean MDI which is used to estimate an overall evaluation of the movement dimension work demand. The steps needed to get to the overall MDI provide information about the specific joints (head/neck, trunk, two shoulders, two elbows, two wrists, two hips, two knees, and two ankles) and body regions (head/neck and trunk, upper limbs, and lower limbs). Thus, this system gives the MDI of each joint and each body region, as well as the overall MDI (Tables 3-1 to 3-4).

Table 3-1. Head/neck and trunk (hnt) movement demand evaluation table: range of motion (ROM), angular motion (AM), joints (j), and movement demand index (MDI).

JOINT	*ROM°*	*AM°*	*MDI_j*
Head/neck*			
Flexion	65		
Extension	50		
Left Lateral Flexion	57		
Right Lateral Flexion	57		
Left Rotation	94		
Right Rotation	94		
Trunk**			
Flexion	85		
Extension	30		
Left Lateral Flexion	28		
Right Lateral Flexion	28		
Left Rotation	38		
Right Rotation	38		
Occupation/Task:	**ROM_{hnt}°:**	**AM_{hnt}°:**	**MDI_{hnt}:**

*Mean ROM values from Nordin and Frankel (1989).

**Mean ROM values from AAOS (1965).

Table 3-2. Upper limbs (ul) movement demand evaluation table: range of motion (ROM), angular motion (AM), joints (j), and movement demand index (MDI).

JOINT	***ROM°***	***AM°***	***MDI_j***
Left Shoulder*			
Flexion	188		
Extension	61		
Abduction	134		
Adduction	48		
Lateral Rotation	34		
Medial Rotation	97		
Right Shoulder*			
Flexion	188		
Extension	61		
Abduction	134		
Adduction	48		
Lateral Rotation	34		
Medial Rotation	97		
Left Elbow*			
Flexion**	52		
Extension**	90		
Right Elbow*			
Flexion**	52		
Extension**	90		
Left Wrist*			
Flexion	90		
Extension	99		
Ulnar Deviation	27		
Radial Deviation	47		
Right Wrist*			
Flexion	90		
Extension	99		
Ulnar Deviation	27		
Radial Deviation	47		
Occupation/Task:	**ROM_{ul}°:**	**AM_{ul}°:**	**MDI_{ul}:**

* Mean ROM values from Chaffin and Andersson, 1991.

** Neutral position = 90° between upper and lower arm.

Table 3-3. Lower limbs (ll) movement demand evaluation table (A = not sitting, B = sitting): range of motion (ROM), angular motion (AM), joints (j), and movement demand index (MDI)

A – Not sitting

JOINT	*ROM°*	*AM°*	*MDI_j*
Left Hip*			
Flexion	113		
Abduction	53		
Adduction	31		
Right Hip*			
Flexion	113		
Abduction	53		
Adduction	31		
Left Knee*			
Flexion	159		
Right Knee*			
Flexion	159		
Left Ankle*			
Dorsi-Flexion	35		
Plantar-Flexion	38		
Right Ankle*			
Dorsi-Flexion	35		
Plantar-Flexion	38		
Occupation/Task:	**ROM_{ll}°:**	**AM_{ll}°:**	**MDI_{ll}:**

B – Sitting

JOINT	*ROM°*	*AM°*	*MDI_j*
Left Hip*			
Flexion**	23		
Extension**	90		
Abduction**	53		
Adduction**	31		
Right Hip*			
Flexion**	23		
Extension**	90		
Abduction**	53		
Adduction**	31		
Left Knee*			
Flexion***	69		
Extension***	90		
Right Knee*			
Flexion***	69		
Extension***	90		
Left Ankle*			
Dorsi-Flexion	35		
Plantar-Flexion	38		
Right Ankle*			
Dorsi-Flexion	35		
Plantar-Flexion	38		
Occupation/Task:	**ROM_{ll}°:**	**AM_{ll}°:**	**PRI_{ll}:**

* Mean ROM values from Chaffin and Andersson 1991.

** Neutral position = 90° between the trunk and the thigh.

*** Neutral position = 90° between thigh and lower leg.

Table 3-4. Overall movement demand evaluation table: movement demand index (MDI).

Body Region	MDI (%)
Head/neck and trunk	
Upper limbs	
Lower limbs (not sitting)	
Lower limbs (sitting)	
OVERALL	

A similar processing of the other relevant variables (repetition, force, and duration) will give a broader representation of the work biomechanical demands, and thus should be conducted.

3.4. Discussion and Conclusions of Chapter 3

A successful ergonomic intervention could result in better control of WMSD. To achieve this major goal of ergonomics, the work physical exertion has to be better understood and the risks need to be systematically classified in a precise and comprehensive manner. The system proposed here may be a first step in this direction. It helps to understand work physical exertion and can be used as a common approach for biomechanical load evaluation. The outcomes of the presented example classify the movement dimension risk of fourteen joints (head/neck, trunk, two shoulders, two elbows, two wrists, two hips, two knees, and two ankles), three main body regions (head/neck and trunk, the upper limbs, the lower limbs), and the overall risk. The use of the system proposed here will show where the physical ergonomic intervention is needed and can also be used to assess the degree of success of the movement dimension load-guided intervention.

The angular motion and position involved in any task influence the amount of force that the worker is able to generate. A possible categorization of the range of motion and position is: (1) within optimal mechanical advantage: within 20% of the neutral position of the joint (Kumar, 1994a); (2) outside the optimal mechanical advantage range: >20% of the neutral position of the joint. The neutral position is defined as the mid range such as 0° between flexion and extension. These parameters can be used to evaluate the angular motion of each joint, showing which joints are most exposed and possibly at risk.

Specifically for the shoulder and back, Punnett, Fine, and Keyserling (1987) found a mean OR of 4.28 for those workers whose AM during work exceed 34% of the ROM.

According to Chaffin (1973) the moment and required muscle force increases 50% when the neck is flexed to 30° (approximately 50% of ROM). Also, the endurance time is significantly reduced with neck flexion ≥30°. Localized fatigue develops with uninterrupted contraction and is associated with localized muscle pain. Neumann *et al.* (2001) studied the relation between physical exposure and low-back pain. Posture and load samples were recorded during the work-shift on a paper using categorical scales. It was found that low-back pain is associated with maximum flexion angle (OR = 2.2), peak spinal loads (OR = 2.0), average spinal loading (OR = 1.7), percent of time with loads in the hands (OR = 1.5), and percent of time spent in flexion > 45°(OR = 1.3) (approximately 50% of ROM as well). In relation to the shoulder joint, Bjelle, Hagberg and Michaelsson (1979) found that close to 70% of the patients at an occupational health clinic with shoulder pain, worked with hands at or above shoulder level. Also, Hagberg and Wegman (1987) found an OR to rotator cuff tendonitis of 11 when working with hands at shoulder level in comparison to work bellow this level.

The duration and repetition of the physical exertion has a cumulative effect on the musculoskeletal system (Kumar 1990). Exertion duration should be taken into account in order to establish work-rest schedules (frequency and duration of the breaks). The proposed system can be applied using sampling techniques or continuous measurement to evaluate the total duration of the work. For jobs characterized by short cycles, direct measures can be taken for the entire cycle and the job does not necessarily need to be broken down into tasks, nor are sampling techniques necessary. For jobs with long cycles, there are still direct measures that can be taken but the job may have to be decomposed into tasks, and those have to be weighted based on the percentage of duration of cycle.

The system proposed here allows flexibility in relation to which methods will be used to collect the biomechanical data and in relation to the number of joints or body regions that will be taken into account. In studies dealing with WMSD, quantitative measures are needed for biomechanical demand analysis of workload. In addition, the effect of ergonomic interventions can be directly evaluated using the classification system proposed. High repetitiveness (0.53 Hz) was found to be directly associated with neck and upper limb disorders; prevalence of 56% on the left hand of female workers (age-adjusted prevalence OR = 3.5) when compared to low repetitive work (0.28 Hz, prevalence of 26%) (Hansson *et al.*, 2000).

A combination of videogrammetry with flexible elgons seems to be the most suitable method of recording the body postures/movements in order to use the proposed classification system. However, in situations that do not allow the use of quantitative tools to directly measure body kinematics, observer based protocols (often used in field studies) may be used. The method proposed here can improve the evaluation of the postural load in both types of assessment (direct and observer based measures). A general assessment of the body posture is ideal for a more complete evaluation. But, in situations where a full recording is not feasible, even the regional assessment of the body region(s) most at risk during the work may add valuable information. Regional assessment is allowed by the proposed system, but single joint assessments should be considered carefully, especially when multiple-joint spanning muscles are involved.

If the task involves the use of only one limb (e.g. use of only the right upper limb to operate a wheel) then the MDI value for that region can be calculated as well. Despite, the evaluator should keep in mind that the general regional indexes gives only a superficial idea of which body regions are mainly involved in the task. However, it is necessary to look at the specific joint indexes in order to identify the risks. If only one joint is been used while the other joints are in a close to neutral position, the regional index may hide and underestimate the risk if used in isolation.

Prolonged and repeated exertions produce harmful physical exposure that can cause musculoskeletal injury, pain, and kinematic disorders. The movement dimension risk can be classified, among other things, according to the percentage of AM in relation to the ROM. The proposed classification system may contribute to a better understanding of the relationship between the occupational biomechanical demands and WMSD. Hopefully, the evaluation system proposed will provide a common approach for occupational biomechanical demand evaluation and help to improve the control of the highly prevalent WMSD. Additional categories of factors involved in the determination of biomechanical demands of work (posture, movement, repetition, force, and duration) can be introduced in order to include other relevant variables, but further studies are necessary to enable the establishment of the appropriate levels and multipliers. Human physiology and biomechanics based elemental behavior and tolerance databases need to be developed against which the industrial load can be compared to assess safety.

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Chapter 4 – Cut-Points for Force Exertion at Work

The need for determining limits for force exertion at work was identified during the formulation of the biomechanical demand evaluation system (Chapter 3). For this reason, I did a literature review on the cut-points for force exertion at work to prevent WLBD. A version of this chapter was published in WORK (Vieira and Kumar, 2006).

4.1. Introduction

Back force exertion at work (e.g. lifting heavy weights) can cause and/or aggravate WLBD (Bernard, 1997). Quantification of the work force demands is needed and it is necessary to "judge" the amounts of force required by the work. Information on the back force requirements of the work can be used to plan and assess interventions to decrease the number of WLBD. Thus, the cut-points are essential to be able to evaluate the appropriateness of the back force demands of jobs because when the demands exceed the workers capacity WLBD may occur (Kumar, 2001; Marras, 2000; McGill, 1997). Cut-points are considered to be the threshold values above which the risk/probability of injury increases significantly. The cut-points reviewed are the limits proposed in the literature for safe back exertion.

The establishment of cut-points for back force exertion during work is difficult due to the great variability of human capabilities and tissue tolerances. Also, the experimental complexity involved in the determination of these limits has imposed restrictions to the current extent of our knowledge in this area. The common approaches used to determine these cut-points employ "consensus" judgments in physiological, biomechanical, and psychophysical criteria. The physiological approach includes measures of variables such as the energy cost of the activities, the oxygen consumption during the job, and EMG of the back muscles involved in the work task. The biomechanical approach includes direct measure of forces exerted by the back muscles to accomplish a job using load cells and/or dynamometers, estimations of compression and shear forces on different back tissues and joints using biomechanical models, measures of joint position (posture) and movement (including range, velocity and acceleration) using different equipment such as elgons and video-recording systems, and measures of back tissues tolerance to stress (mainly *in vitro* experimentation). Finally, the psychophysical approach involves worker/subject definition of acceptable levels of activity. In this case, as stated in Snook and Ciriello (1991), workers are instructed to "work as hard as you can without straining yourself, or without becoming unusually tired, weakened, overheated, or out of breath". The psychophysical approach is used to define activity levels that are acceptable to different percentages of the population (e.g. 10, 25, 50, 75, and 90% of the female and male population).

Despite the availability of cut-points for back force exertion at work, there are no gold standards and each of the proposed cut-points has weaknesses according to the assumptions made in establishing each of the proposed values. Also, the available recommendations are not presented concisely in the scientific literature making the decision making process difficult. For these reasons, the objective of this paper was three fold – (I) to critically review the scientific literature concerning the cut-points for force exertion using the back during work; (II) to present the available guidelines in a concise manner; and (III) to identify areas that need further research.

4.2. Methods

This review included articles that met the inclusion criteria stated below. Papers published in the English language before 2004 including different combinations of the following words in the title, abstract, or keywords were searched in the PubMed (Medline), Scirus, and Science Direct (Elsevier Science) databases – "force", "work", "low back", "exertion", "cut-points", "threshold values", "limits", and "tissue tolerance". To be included, the articles had to address back force exertion cut-points, include a classification of the cut-point (e.g. increased risk of WLBD if exertion is above the cut-point), use quantitative measures (load cell, dynamometer, EMG, biomechanical models), present objective results (e.g. OR, RR, injury prevalence), and not be based only on expert opinion. Relevant papers in the reference lists of those articles selected as well as review papers connecting the results of articles meeting the established criteria and pieces of information from relevant textbooks were also included. The studies reviewed were grouped based on the criteria used to set the cut-point values in "weight and distance of the load from the body", "percentage of maximum voluntary contraction used", "acceptable loads based in worker opinion", "weight and number of repetitions", "intra-abdominal pressure", "spinal compression forces", and "compound index".

4.3. Results

Different approaches have been used to propose cut-points for back force exertion. Table 4-1 presents a summary of the studies reviewed by criteria used including references, proposed cut-point, classification of the cut-point, and possible problems.

Table 4-1. Summary of the studies including cut-points for back force exertion.

Criteria	References	Cut-point(s)	Classification(s)	Problem(s)
Weight and distance of the load from the body	Chaffin and Park (1973)	Weight > 16 kg; Weight > 9 kg if horizontal distance > 51 cm; Max. weight / max. capacity of "strong man" ratio >0.2.	Increased risk of WLBD*. Recommendations for lifting, based on incidence rates (n = 411)	Only applicable for sagittal lifting. The lifting strength rating uses a pre-defined maximum capacity not considering workers variability and duration of force exertion.
Percentage of maximum voluntary contraction (MVC) used	Lindstrom *et al.* (1977)	Contraction ≥ 15 ≤ 20% of MVC. Contraction ≥ 60% of MVC.	Decrease in blood flow. Blocked blood flow, fatigue.	The values depend on the musculature evaluated (percentage of different type of muscle fibers).
	Sjøgaard (1986) and Sjøgaard *et al.* (1986)	Contraction of 5% of MVC for 1h.	Lower membrane potential and alteration of muscle excitability.	
Acceptable loads based in worker opinion	Snook and Ciriello (1991)	Weight > acceptable to 75% of female and 99% of male workers. 23 kg for lifting occasionally in the sagittal plane, good couplings, and vertical displacement ≤ 25 cm.	Increased risk of lifting-related WLBD. Maximum recommended weight.	Tolerance levels of workers do not represent a limit for injury precipitation.
Weight and number of repetitions	Ahlborg *et al.* (1990)	Weight ≥ 12 kg for > 50 times/week.	Increased risk of pre-term birth Odds Ratio (OR) of 1.7.	
	Punnett *et al.* (1991)	Weight ≥ 4.54 kg when repetition > 1 time/min. during the entire workday.	Increased risk of WLBD OR 2.2.	The samples back strength level may be higher than same working populations.
	Hoogendoorn *et al.* (2000)	Weight ≥ 25 kg for >15 times/workday.	Increased risk of WLBD Relative risk (RR) of 1.79.	
Intra-abdominal pressure (IAP)	Davis and Stubbs (1977a, b and 1978)	≥ 90 mmHg	Increased risk of WLBD	The effects of increased IAP are controversial

* WLBD – work-related low back disorders.

Table 4-2. Summary of the studies including cut-points for back force exertion (continued).

Criteria	References	Cut-point(s)	Classification(s)	Problem(s)
Spinal compression force	Evans and Lissner (1959) and Sonoda (1962)	3400 N 6700 N	Micro-fractures of vertebral endplates: ≥ 60 years old. < 40 years old.	
	Chaffin, Park (1973)	2500 N 4500 N	5% WLBD incidence rates. 10% WLBD incidence rates.	
	NIOSH (1981)	3400 N 6700 N	Action limit (AL) for compression at L5/S1. If > AL, increased risk of WLBD. Maximum permissible limit.	
	Hutton and Adams (1982)	10249 N	Ultimate compressive axial force of intervertebral discs of cadavers of males between 22 and 46 years old.	
	Adams and Hutton (1982)	5400 N	>40% of the intervertebral disks prolapsed. Flexed spines: simulated by wedging vertebral bodies.	The compression values are means with high standard deviations. The data is from the study of cadavers, the behavior of living structures might differ. The cut-points proposed do not consider cumulative effect, and are based on axial compression only.
	Adams and Hutton (1985)	3800 N	Intervertebral disks trabecular fracture after repetitive loading of simulated flexed spines.	
	Herrin *et al.* (1986)	≥ 4500 N	WLBD IR 1.5 higher than when compression < 4500 N.	
	Brinckmann *et al.* (1987), Biggemann *et al.* (1988), and Brinckmann *et al.* (1989a,b)	Loads ~ 60% of UCS**; 5000x; over 6h. Loads ~70% of UCS; 500x; 30min. Loads ~75%UCS; 10x; 40sec.	Increased risk of fracture. >90% of lumbar vertebrae fracture	
	Jager and Luttmann (1989)	4400 N	Axial compression limit.	
	Norman *et al.* (1998)	3423 N	Increased risk of WLBD: OR = 1.9	
Compound index	NTIS (1991) and Waters *et al.* (1993)	Lifting index > 1. Lifting index > 3.	Increased risk of WLBD "for some workers". Elevated risk of WLBD "for many workers".	There are several restrictions to the use of the method (e.g. it requires use of both hands, specific ranges of motion, velocity, etc) and it is limited to lifting only.

* WLBD – work-related low back disorders.

** UCS – ultimate compressive strength.

4.3.1. Weight and distance of the load from the body

Chaffin and Park (1973) conducted a one-year longitudinal study about the relationship between WLBD and occupational lifting. Their study included 411 subjects from 103 jobs. They suggested that not only the load has to be taken into account but also the distance between the load and the body. Based on the results, they stated that there is increased risk of WLBD when workers lift more then approximately 16 kg, and if the horizontal distance of the load is more than approximately 51 cm from the ankle, lifting even lower loads (more than approximately 9 kg) already represents increased risk of WLBD.

Chaffin and Park (1973) normalized the data by dividing the maximum weight lifted by the maximum lifting capacity of a "large/strong man". The authors called this method "lifting strength rating (LSR)". Based on their findings, they stated that when the load is above 20% of the maximum lifting capacity of a "large/strong man" (LSR >0.2) the worker is at increased risk of WLBD. According to the authors, 0.2 LSR weight is close to what 95% of women are capable of lifting close to the body (see section 4.3.3 "Acceptable Loads Based on Worker Opinion"). The limitation of this study is that the considerations are made for lifting with both hands in the sagittal plane. This ideal situation is not always present in the work environment. This way, lower values may represent risk to the musculoskeletal system when the conditions are different from those studied (during asymmetric lifting, for example).

Warwick *et al.* (1980) found a reduction of 38% to 50% of the maximum voluntary strength with increasing lifting asymmetry in comparison with sagittal plane lifting. Weight and distance of the load from the body is a parameter frequently used to assess workload. These parameters are directly related to the resulting spinal compression forces. They are used on the biomechanical models and compound indexes, and are further discussed in Sections 4.3.6 and 4.3.7 of this paper.

4.3.2. Percentage of maximum voluntary contraction (MVC) used

Localized fatigue develops with uninterrupted contraction and is associated with localized muscle pain. A study by Lindstrom *et al.* (1977) presented a method for evaluation of muscle fatigue by power spectrum analysis of EMG signals. The results showed that during contractions from 15% to 20% of MVC blood flow starts to be impaired; at 60% of MVC blood flow is totally blocked. Decreased or interrupted blood flow is related to muscular fatigue due to intracellular acidosis (accumulation of metabolites) and/or lack of energy (lack of substrate supply).

Even low levels of muscular contraction can cause fatigue by decreasing potassium concentration in the muscle resulting in muscle fibers' excitability alteration. For example, Sjogaard (1986) and Sjogaard *et al.* (1986) found a lower membrane potential after 1 h contraction at 5% of MVC even though the blood flow was not affected; there was a 12% reduction in MVC for the knee-extensors. Thus, independently of the level of contraction, rest periods are necessary for recovery since no contraction can be maintained continuously. Safe levels of continuous low level isometric contractions need to be determined to establish guidelines for work-rest schedules.

4.3.3. Acceptable loads based in worker opinion

Snook and Ciriello (1991) published several tables presenting psychophysically determined maximum acceptable weights and forces for lifting, lowering, pushing, pulling, and carrying tasks. The tasks varied in frequency, distance, height, and duration. The objects varied in size and design (boxes with and without handles). Due to the extensive data presented by the authors the entire list of recommendations is not included in this paper; for additional information reefer directly to their paper. The study by Snook and Ciriello (1991) was used to generate the psychophysical criteria used in the NIOSH 1991 lifting equation to establish 23 kg as the maximum recommended weight in "optimal conditions" (occasional lifting in the sagittal plane, with good couplings, and vertical displacement of less than 25 cm). The proposed cut-point is associated with approximately 3400 N of spinal compression, 3.5 Kcal per minute energy expenditure, and is acceptable by 99% males and 75% females; but it increased the risk of WLBD moderately (Waters *et al.*, 1993).

The 23 kg cut-point for lifting proposed is the revised limit proposed by NIOSH; the initial limit was much higher (40 kg) (Konz, 1982). Even though, NIOSH guidelines take into consideration some cadaver studies about intervertebral disc tolerances, the recommendations are not derived from injury causation data. The new recommendation is based on psychophysical determination of acceptable loads to be lifted for different durations and frequencies. Psychophysical studies may show the tolerance levels of workers but do not represent a limit for injury precipitation. The same psychophysical studies by Snook and Ciriello were used to define the limits for pushing, pulling and carrying; consequently they present the same the same limitations. Relying on a worker's perception of the biomechanical stresses may not be adequate to protect the person (Freivalds *et al.*, 1984; Kumar and Mital, 1992).

Studies have shown that workers perception of the amount of force being exerted is inaccurate (Strindberg and Petersson, 1972; Wiktorin *et al.*, 1996). On the other hand, Snook (1978) reported that the group of workers who performed activities that were accepted by less than 75% of all workers presented three times more WLBD than those whose activities were accepted by more than 75% of the working population. NIOSH guidelines for back force exertion at work are further discussed in the "Spinal Compression Forces" section of this paper.

Psychophysical studies also have shown differences in maximal acceptable weight (MAW) for non-sagittal lifting. Ljungberg *et al.* (1982) reported that for lateral transference (e.g. horizontal lifting from one table to another) subjects chose weights of only half of the load compared to the weight reported in other studies as chosen for lifting in the sagittal plane. Garg and Badger (1986) conducted another psychophysical study on MAW and maximum isometric strengths (MIS) during symmetric (0° in relation to the sagittal plane) and asymmetric lifting (30°, 60°, and 90°) at a frequency of one lift every five minutes. Both MAW and MIS decreased by increasing the degree of asymmetry ($p < 0.01$). There was 23% reduction in MAW for asymmetric lifting at 90°, 15% at 60°, and 7% at 30°. In relation to MIS, there was 31% decrease at 90°, 21% at 60°, and 12% at 30°. The fact that the percentage decrease in MIS was higher than the percentage decrease in MAW for the three positions, supports the previously discussed issue that the tolerance levels of workers may not represent a limit for injury precipitation since the percentage of MIS is increasing without being accounted for or noticed by the workers. The authors suggested the following equation for the calculation of the MAW (EQ 2).

Calculation of the maximal acceptable weight (MAW):

$$MAW = 18.1 + (0.528MIS) \qquad \text{(EQ 2)}$$

Where: MAW = maximum acceptable weight in kg; MIS=maximum isometric strength [adapted from Garg and Badger (1986)].

One of the limitations of these calculation is that the formula is based on a sample of thirteen healthy young male subjects (age range from 21 to 35 years old), with a determination coefficient (r^2) of 0.62 and a standard error of 5.6 kg. The MIS is highly dependent upon posture and presents high coefficients of variation. The average MAW was approximately 38 kg which is 65% higher than the cut-point for lifting proposed by NIOSH (23 kg).

4.3.4. Weight and number of repetitions

In addition to NIOSH and other cut-points discussed previously, this section presents additional information. In a three-year prospective cohort study including 861 workers, back force exertion was critical to the development and aggravation of WLBD (Hoogendoorn *et al.*, 2000). For the sub-group of 724 workers with no or minor changes in work during the three-year period, those who lifted weights of at least 25 kg more than 15 times/workday were found to be at higher risk of having WLBD (RR = 1.79). Lifting is a risk factor not only for WLBD; in a prospective study done in Sweden involving 3906 pregnant workers, lifting 12 kg or more (50 or more times per week) was found to be a risk factor (OR 1.7) for pre-term birth (less than 37 weeks of gestation) (Ahlborg *et al.*, 1990).

4.3.5. Intra-abdominal pressure

Early studies attributed reduction of intervertebral disc pressure during lifting to increased intra-abdominal pressure (IAP) (Bartelink, 1957). In 1977, Davis *et al.* published a paper about the use of radio pills to monitor back stress. After being swallowed by the subjects, these radio pills were able to measure IAP. This pressure is related to the amount of contraction of the abdominal muscles and it was believed that IAP was linearly related to the compression forces acting on the spine. Based on these principles the authors performed several studies about safe levels of manual forces for young males (Davis and Stubbs, 1977a, b, and 1978). The main recommendation from these studies was that IAP should not exceed 90 mmHg. This value was derived based on the fact that the incidence rates of WLBD increased significantly when pressures of 100 mmHg and higher were present in specific occupations.

The reduction of intervertebral discs pressure by increasing IAP was later refuted. Kumar (1980) found that the IAP increase was concurrent with increase in the back muscles EMG. Later, it was found that a Valsalva maneuver ("voluntary pressurization of the intra-abdominal cavity") increased intervertebral disc pressure in the upright standing posture, even though the maneuver decreased the intervertebral disc pressure in a stooping subject (Nachemson *et al.*, 1986). Additionally, the temporal recording of IAP along with EMG demonstrated that IAP was neither related to force nor EMG (Kumar, 1997). Instead it was found to be a byproduct of other physiological phenomena with no consistent relationship with disc compression. The effects of increased IAP are controversial and it can not be assumed to always reduce intervertebral disc pressure. Even if it were to reduce disc compression, the impact of IAP would be too small to be of material relevance (Steven Wiker in personal communication).

4.3.6. Spinal compression forces

Chaffin and Park (1973) found WLBD incidence rates of 5% and 10% for workers ($n = 411$) exposed to estimated compressive force at L5/S1 higher than 2500 N and 4500 N, respectively. Another study showed that L5/S1 intervertebral disk compression is a good predictor of back and other overexertion injuries (Herrin *et al.*, 1986). For jobs with estimated compressive force at L5/S1 between 4500N and 6800 N, the authors found a rate of WLBD more than 1.5 times higher than for jobs with estimated compressive force lower than 4500 N (revised results after Waters *et al.*, 1993). In addition, cumulative load (compression and shear over an individual's working life) was shown to be higher in subjects with back pain (Kumar, 1990).

NIOSH (1981) proposed guidelines for the assessment of manual lifting (these guidelines were later revised and are discussed later in this chapter). The 1981 guidelines defined an action limit (AL) and the maximum permissible limit (MPL = 3AL). According to NIOSH, an ideal work environment should keep the exposure under or close to the AL and never exceed the MPL. NIOSH suggests a MPL for compression at L5/S1 intervertebral disk of 6700 N, and an AL value of 3400 N (NIOSH, 1981).

NIOSH guidelines for compression are based on the studies of Evans and Lisner (1959), and Sonoda (1962). The results of these studies show that, even though the intervertebral discs do not rupture, micro-fractures of the vertebral cartilage endplates (spines from cadavers of subjects under 40 years old) start to happen when applying on average 6700 N of axial load (approximately 680 kg). When the spines were from cadavers 60 or more years old, micro-fractures appeared when applying average axial loads of 3400 N. Based on these findings, NIOSH suggested a maximum acceptable compression at L5/S1 intervertebral disk of 6700 N for subjects under 40 years old, and 3400 N for subjects with age of 60 or more.

In addition to the studies used to define NIOSH 1981 guidelines, the results of some studies performed after 1981 have supported the initial recommendations. Jager and Luttman (1989) compared the results from their proposed biomechanical model for low back axial compression estimation with the literature regarding lumbar compressive strength. The average ultimate axial compression strength (total of 307 lumbar segments) reported by the authors was 4400 N (SD = 1900).

Norman *et al.* (1998) compared a 104 cases (with WLBD) with 130 controls (without WLBD), predicted peak shear force on L4/L5 (OR = 2.3) and peak compression force on L4/L5 (OR = 1.9) emerged as risk factors for WLBD. The mean peak compression load of the auto-assembly workers who reported WLBD was 3423 N. This value is approximately the same as the AL (3400 N) proposed by NIOSH (1981) and it was significantly different ($p < 0.001$) form the mean value found for the group who did not report WLBD (2733 N).

Hutton and Adams (1982) found a mean ultimate compressive axial force of intervertebral discs of cadavers of males between 22 and 46 years old value of 10249 N. However, more than 40% of the intervertebral disks prolapsed when 5400 N of axial load was applied to flexed spines (simulated by wedging vertebral bodies). Additionally, in another study the authors observed trabecular fractures in the intervertebral discs when an average repetitive axial load of 3800 N was applied to hyper-flexed spines (Adams and Hutton, 1985).

Lumbar vertebral fracture due to axial compression has also been studied. In the studies of Brinckmann and colleagues, repetitive loads from 20% to 75% of the ultimate compressive strength were applied to cadaver spinal segments at a frequency of 15 repetitions/minute, up to 5000 times (maximum time of approximately 6h) (Biggemann *et al.*, 1988, Brinckmann *et al.*, 1989a and b). The ultimate compressive strength was predicted from the area of the vertebral end plate and from the trabecular bone density measured by quantitative computed tomography. Ninety-two percent of lumbar vertebral specimens suffered fractures when loads of 50 to 60% of the ultimate compressive strength were applied 5000 times over approximately 6h. But, when the percentage of the ultimate compressive strength was increased by only 10%, 91% of the vertebrae suffered fracture after 500 repetitions in approximately half an hour of testing. For loads of 75% of the ultimate compressive strength, the vertebrae suffered fracture after only 10 repetitions (40 seconds of testing).

The major limitation of NIOSH 1981 guidelines is that the cut-points are based on cadaver studies with large standard deviations, and the living structures threshold to compression injury for different people might differ. Waters *et al.* (1993) question these values, specially the AL value of 3400 N. NIOSH's opinion is that this AL value "may not protect the entire workforce" (Waters *et al.*, 1993). In addition, the guidelines are based on studies of axial compression only and do not take into account the cumulative effect and temporal characteristics of the exertions over time on the viscoelastic tissues of the body (Kumar, 1990; Van Dieen and Vrielink, 1994). The compression guidelines proposed by NIOSH are widely used, however, as suggested by different studies, they are probably inaccurate and when followed may expose the workforce to demands exceeding its capacity.

4.3.7. Compound index

In addition to the cut-points mentioned separately for each variable, specifically for the low back, the NIOSH 1991 lifting equation provides a compound measure (lifting index) to assess the risk of lifting-related WLBD (Waters *et al.*, 1993). The NIOSH lifting equation and its cut-points are used to assess the risk associated with lifting in many places in addition to the U.S. such as Canada and European countries (e.g. Health

Council of the Netherlands, 1995). The lifting index takes several variables into account to calculate the risk of WLBD including object weight, position, hands coupling, vertical and horizontal displacement, posture, L5/S1 compression force, frequency, and duration. The lifting index is a ratio between the actual weight lifted (AWL) and the "recommended weight limit (RWL)" calculated with the following equation (EQ 3).

Recommended weight limit (RWL) calculation:

$$RWL = 23 \times \left[\frac{25}{h}\right] \times \left[1 - 0.003\left|v - 75\right|\right] \times \left[0.82 + \frac{4.5}{d}\right] \times f \times \left[1 - 0.0032 \times a\right] \times c \quad \text{(EQ 3)}$$

Where: RWL = recommended weight limit in kg; h = horizontal distance of the hands in relation to the ankles in cm; v = vertical distance of the hands in relation to the floor in cm; d = vertical dislocation of the weight in cm; f=frequency multiplier from a table; a = angular dislocation of the weight; c = coupling multiplier from a table [adapted from Dempsey (2002) and Waters (1993)].

The lifting index method (AWL divided by RWL) suggests the following cut-points: when the lifting index is above 1 there is increased risk of WLBD "for some fraction of the workforce"; when the index is above 3, there is elevated risk of WLBD "for many workers". The main limitations of this index is that it only applies for lifting tasks and does not take into account other tasks that are frequently related to lifting jobs such as pushing, pulling, and carrying. Dempsey (2002) studied "primarily lifting/lowering jobs" including 449 workers. Approximately 56% of he workers performed pushing, pulling, and/or carrying tasks in addition to lifting and lowering.

Even for lifting tasks there are several restrictions to the use of the lifting index. For example, the lifting equation requires use of both hands, specific ranges of motion, velocity, repetitions per minute, duration, and unrestricted work space. van der Beek *et al.* (2000) analyzed 559 lifting tasks and reported that 57% of the tasks could not be evaluated using the NIOSH lifting equation. In addition, studies have shown that the lifting index tends to overestimate the risk of WLBD; in a study by Waters *et al.* (1998), only one (lifting index = 2.7) out of fifteen analyzed lifting tasks presented a lifting index less than 3. A problem with the lifting and severity indexes is that some users incorrectly consider the ratio as an index of spinal injury risk. The RWL and the AL fluctuate as a function of the biomechanical, physiological, psychophysical, and other consensus criteria. In addition, the LI ratios for different lifts can be equal for very different reasons, or produce equivalent biomechanical stress yet have very different ratios. The LI should represent a lifting task design quality metric as opposed to a injury risk index (Steven Wiker in personal communication). Another limitation is that many of the individual cut-points used to derive the compound measure are based on psychophysical data (the limitations of this approach were discussed earlier in this paper). Further studies are necessary to test the validity of the cut-points of the lifting index.

4.4. Discussion and Conclusions of Chapter 4

Physical exertion at work is often analyzed using observational methods including nominal classification (light, intermediate, heavy) and/or interval scales (< 5 kg, 5–10 kg, >10 kg). Due to lack of precision in the observational methods and the current availability of direct measures, the use of quantitative devices (continuous scale data) instead of observation methods is encouraged (Vieira and Kumar, 2004b). This need is further exemplified by the following quotes: "direct measurement methods are the only serious option to assess the level of the exerted forces with the accuracy needed for ergonomic epidemiology" (Van der Beek *et al.*, 1999); and "direct observation can be used to capture the postural and temporal demands of work, but quantitative assessment of force is only possible through direct measurement" (McGill, 1997). However, caution is needed when using direct measures because the devices may affect the task. Calibration and normalization procedures are also necessary. In addition, suitable instruments need to be developed and/or improved for onsite data collection.

Despite the type of method used to collect the data, the amounts assessed need to be evaluated (Vieira and Kumar, 2004a). Frequently, the success of ergonomic interventions is based on the reduction of the work biomechanical demands. Usually it is accepted that "the less the better"; however, these statements are not adequate. We need to start being more critical in our evaluations in order to succeed in reducing the WLBD. We should evaluate the magnitude of present exertion instead of accepting "lower values" after an intervention as a positive outcome.

In addition, the epidemiological studies could be designed such that groups of workers exerting specific amounts of back force during the job were compared instead of just comparing exposed and non-exposed workers. This approach would possibly help to determine valid cut-points for back force exertion. So far, the studies are designed to study the relationship between higher exposure and higher WLBD incidence instead of testing specific amounts of back force exertion as possible cut-points.

Peak load decreases may not reduce injury risk if cumulative load is increased (Daynard *et al.*, 2001; Kumar, 1990; Norman *et al.*, 1998). However, no clear cut-points for cumulative back force exertion were found. Further experimental and epidemiological studies in peak load and cumulative exposure are necessary. Often there is job rotation in industrial workplaces, cut-points for safe back exertion for shorter exposures than 8h are also necessary. However, usually the workday is still 8h or so, thus the 8h cut-points for each of the jobs in the rotation scheme may further protect the workers against WLBD by indirectly taking into account the cumulative effect of physical exertions. There will not be a specific cut-point value that will be applicable to each and every situation because the characteristics of both the working population [e.g. cut-points for pregnant workers (Morrissey, 1998); cut-points for older workers (Kumashiro, 2000)] and job task can differ significantly [e.g. cut-points for peak vs. cumulative exertion (Daynard *et al.*, 2001)]. However, cut-points are necessary as normative information and should be as specific as possible. Another consideration is in relation to the validity and utility of the available cut-points.

Even though some studies have shown higher incidence of WLBD in jobs with higher compressive force at L5/S1 (e.g. Herrin *et al.*, 1986), it does not mean that the compression is causing the injuries. Compression may not be the causal factor of WLBD, but it may be occurring in jobs that also have other factors such as repetition, trunk rotation, and awkward postures that increase the load on back muscles and ligaments. Actually, it is already known that axial compression is not the main cause of intervertebral disc herniation. Even more importantly, less than 20% of WLBD are discogenic. Notwithstanding, cut-points for L5/S1 compressions are often used as safety guidelines for manual materials handling tasks. More attention should be paid to the erector spinae muscle because it is the main trunk extensor (Macintosh and Bogduk, 1991). The use of EMG recognition of muscle fatigue has potential and interesting applications for WLBD prevention. Further research is necessary to determine cut-points for back force exertion based on EMG determined muscle fatigue.

Muscle strength is proportional to its cross-sectional area (CSA). EMG fatigue and CSA are used in biomechanical models to predict muscle forces considering the muscles fascicle orientation (Macintosh and Bogduk, 1991; McGill, 1992). Jorgensen *et al.* (2003) measured CSA of lumbar spine muscles in different sagittal postures using MRI. The mean maximum CSA in the neutral posture was 23.7 cm^2 (SD = 3.5) for males (n = 12), and 14.8 cm^2 (SD = 2.0) for females (n = 12). The authors found that the CSA of lower lumbar spine muscles changes with trunk motion, but the maximum CSA of lumbar spine muscles is not affected by trunk posture in the sagittal plane. Gender, body mass, torso area, and spinal curvature can be used to predict lumbar muscles CSA. Considering that the force generation capability of 1 cm^2 of muscle is approximately 35 N on average (Kroemer and Grandjean, 1999), it can be estimated that, on average, young healthy male and female subjects are able to generate maximal forces with the lumbar spine muscles of approximately 830 N (35 $\times$ 23.7) and 520 N (35 $\times$ 14.8), respectively.

The maximum weight that could be lifted using the back can be calculated using biomechanical models considering the distance of the weight from the body and the trunk posture. However, these are estimates of the maximum strength capabilities and there might be WLBD risk even when only 20% of MVC is used (Lindstrom *et al.*, 1977). The use of data regarding the CSA of the back muscles exerting the forces is needed to define the percentiles of the population that the cut-point is reliable. For this reason, as previously discussed by Jaric *et al.* (2002), normalized instead of absolute strength measures are necessary and require further research. As commented in Kumar's Annual Ergonomics Society Lecture in 2003, ". . . the adjustments made in standards so far do not adequately meet the human limitation" (Kumar, 2004). There are cut-points in the literature for back force exertion, but definitive cut-points were not found. Actually, according to Kroemer (1999) "... regarding muscle functions, much research was directed at the isometric condition and, consequently, most information available on muscle strength concerns this static case. Data on human body strength are still largely limited to static or quasi-static conditions". Most often the available cut-points differ than concur. Further studies are necessary and should quantitatively address the level of both isolated and combined back exposure. The NIOSH approach in developing the 1991 lifting equation considering physiological, psychophysical, epidemiological, and biomechanical aspects of exertion (NTIS, 1991) meets the most known criteria and present the lowest common denominator for lifting tasks.

The reference values being used to-date do not seem to be optimally effective. Evidence of this inadequacy is given by the limited success achieved so far in controlling WLBD. In addition, compound indices should also be developed for other types of back force exertion that are common place at work such as pushing, pulling, and carrying. These indices should consider EMG determined fatigue, differential viscoelastic properties of tissues, aging, and the cross sectional area of back muscles. We hope that this paper contributes to a more systematic appraisal of back force exertion at work.

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Chapter 5 – Estimating Work-Related Load on the Lumbar Spine with Biomechanical Models: a literature review

During the review of the cut-points for force exertion at work the important role of biomechanical models was identified. There is no "gold standard" model to estimate spinal loads and different studies use different methodologies. For these reasons, my colleague Catherine Trask (University of British Columbia – UBC) and I, with the support of our supervisors: Dr. Shrawan Kumar (University of Alberta), Dr. Mieke Koehoorn (University of British Columbia) and Dr. Kay Teschke (University of British Columbia) decided to do a literature review on biomechanical models to estimate low back loads in occupational settings. This chapter presents the literature review which we are improving to submit for publication.

5.1. Introduction

Work-related low back disorders (WLBD) are the most frequent and most costly musculoskeletal disorder, accounting for 20% to 30% of all work-related compensation claims in Canada (Statistics Canada, 1995; AHRE, 2003) and for approximately 40% of all lost time claims in the USA (Guo *et al.*, 1999). To address this problem, assessments of WLBD risk factors are needed to classify jobs and tasks, to identify needs for intervention, and to evaluate its effectiveness. Biomechanical loads on the spine are recognized risk factors for WLBD (Bernard, 1997). Giovanni Alfonso Borelli (1608-1679), considered the "father of biomechanics", was the first to combine mathematics, physics, and anatomy to describe the movement of the body in his seminal work, *du motu animalium* (Provencher, 2000; Sanan, 1996). According to Kumar (2004), the first biomechanical model of the spine was presented by Morris *et al.* in 1961. In 1969, Chaffin presented a model to estimate spinal load resulting from holding an external static load in a specific sagittal posture using a single back extensor muscle equivalent (Chaffin, 1969), with many subsequent improvements. Shultz and Andersson (1981) included multiple muscle torques in a biomechanical model to estimate lumbosacral compression and shear. Subsequent models increased the number of muscles in the calculations and optimization strategies (e.g. Bean *et al.*, 1988; Cheng and Kumar, 1991; McGill and Norman, 1986; Schultz *et al.*, 1982). Sophistication was further increased by the introduction of three-dimensional and EMG-assisted models and by stochastic or probability-based estimates of spinal loads (Cholewicki and McGill, 1994; Marras and Somerich, 1991; McGill and Norman, 1985; Mirka and Marras, 1993).

There are a number of biomechanical models available to estimate occupational loads on the lumbar spine. Each model has strengths and weaknesses according to its assumptions. There is no agreement on a "gold standard" model to be used in workplace settings. Also, the different models available are not presented concisely in the scientific literature, making it difficult to select the most appropriate model for specific purposes. This paper discusses the strengths, weaknesses, assumptions, and limitations of biomechanical models to estimate spinal loading during occupational activities in workplace settings. In this review, biomechanical models were classified as static, quasi-dynamic, or dynamic, and were discussed in relation to their reliability, validity, and practicability. Thus, the objectives of this paper are to: a) critically review the scientific literature on the estimation of loads on the lumbar spine in the workplace using biomechanical models, b) to present the available models and methods in a concise manner, and c) to identify areas that need further research. This review is intended to help practicing professionals in occupational biomechanics and ergonomics in selecting the most appropriate biomechanical model for specific purposes.

5.2. Methods

Two separate searches for English-language scientific papers in the PubMed and Science Direct databases were carried out. The searches were performed for all available years (up to Oct 2005) using the same set of keywords on each database. The keywords (Table 1) were searched within the title, abstract, and keyword fields on Science Direct, and with the titles and abstract fields on PubMed. The keywords were categorized into three subject domains ("low back", "work-related", and "biomechanical model") with synonymous or conceptually similar terms separated by the bouillon connector "OR" in order to maximize the search outputs. The three subject domains were then combined using the bouillon connector "AND".

The resulting titles from the independent searches were merged into one database and the duplicates were deleted. Subsequently, the two authors independently reviewed the titles for exclusion of non-relevant papers based on the exclusion criteria presented in Table 10. This was followed by a thorough review of the abstracts, again conducted by each author independently.

The papers that met the inclusion criteria (also presented in Table 5-1) after the abstract review were selected. All papers chosen were reviewed. A review of the bibliography of the selected papers was undertaken to identify additional papers.

Table 5-1. Keywords searched, inclusion criteria, and exclusion criteria used to define the papers included in this review of the literature on the estimation of loads on the lumbar spine in the workplace using biomechanical models.

Keywords Searched	**Exclusion Criteria**	**Inclusion Criteria**
-low back OR -lumbar OR -spinal OR -low back disorders OR low back pain AND -work-related OR workload OR industrial OR occupational OR physical demands AND -biomechanical model OR biomechanical -OR biomechanics	Titles clearly indicating that the paper was about: -spinal cord injury; -cervical or upper limb disorders; -functional capacity evaluation; -spinal posture measurement unrelated to occupational situations; -athletic performance; -models of whole body vibration; -biomechanical models only suitable for laboratory studies; -studies that used the models without evaluating them	Studies presenting or evaluating: -biomechanical models to estimate low back workload; -biomechanical models suitable for field studies of occupational demands

Included articles described models that were either intended to assess occupational loads in a workplace setting or employed methods that were conducive to this purpose. Models were required to allow the assessment of job tasks without significantly restricting the worker or interfering with the tasks. Considerations were made on the amount and complexity of instrumentation; for example, models involving more than 8 channels of EMG or extensive set-up and calibration procedures were deemed impractical for field use. For the purposes of this review, we have divided the biomechanical models to estimate occupational loads on the lumbar spine into three categories: static models which assume static equilibrium loading conditions, dynamic models which account for inertia associated with movement of the external load and body segments, and quasi-dynamic models which combine some elements of both static and dynamic models. This classification was chosen because it incorporates the theoretical composition and it has implications to the appropriate application of the different models.

5.3. Results

5.3.1. Search and selection outcome

The initial keyword search yielded 320 titles on Science Direct and 506 titles on PubMed. Of these, 312 were duplicate titles between the two searches, yielding a database of 514 (312+8+194) titles. After individually reviewing the titles, the two authors excluded 153 and 172 titles respectively; 105 of the excluded titles were duplicates. The 115 (48+67) titles excluded by only one reviewing author were discussed resulting in 90 exclusions. In total, 195 (105+90) titles were excluded after title review. Of the 319 (514-195) abstracts reviewed, the authors selected 68 and 76 abstracts respectively; 48 of these were duplicates. The abstracts selected by at least one of the authors were included. The number of papers selected for critical review was 96 (48+20+28).

5.3.2. General features of biomechanical models

Occupational biomechanical models of the spine must include the relevant loading variables (e.g. lifted or carried loads, distance between the external load and the body, pushing and pulling forces, trunk flexion, rotation, and lateral bending angles) as inputs in order to estimate loads on the low back. Alternatively, the loads may be estimated from calibrated surface EMG. The biomechanical risk factors for WLBD may be grouped as kinematic or kinetic. Kinematic aspects include the body motion, velocity, and acceleration, while the kinetic aspects include the moments, loads, compression and shear forces. The biomechanical loading parameters can be assessed using motion analysis, electromyography, load cells, dynamometers, and force platforms (Vieira and Kumar, 2004; 2006).

The biomechanical models may use direct or inverse dynamics to estimate moments, loads, and strength requirements. Direct dynamics uses the information about a load's distribution and accumulation starting from an external load measured in the hand, upper limbs, and trunk to the spine and lower limbs (downward approach), while inverse dynamics uses the ground reaction forces from force platforms or insoles to arrive at the loads acting on the spine and other joints (upward approach) (Magnusson, 1990; van der Beek, 1998). For the latter, the reaction forces due to the weight of the lower limbs and any contact forces below the spine level should ideally be discounted from the calculations.

Some biomechanical models attempt to partition the load among different tissues, but most models estimate net moments, compressions, and shear forces in the lumbar spine at L4/L5 or at L5/S1, with compression being the most common load parameter. Spinal compression has long been studied as a risk factor for WLBD, and forms the basis of the NIOSH lifting recommendations (Waters *et al.*, 1993). Compression and shear forces are both affected by trunk motion, muscle contraction, and external loads. Norman *et al.* (1998) found a strong correlation (r = 0.83) between compression and shear forces. Meanwhile, Marras and Somerich (1991b) reported a trade-off between compression and shear, whereby decreased compression and increased shear loading occur under asymmetrical conditions.

Despite the frequent use of spinal compression estimates from biomechanical models, their preeminence as risk factors for WLBD has been questioned. According to Kumar (2004), "even though we know that more than 80% of all back disorders are not discogenic, we continue to model for disc compression values". Granata and Marras (1999) suggested that spinal compression is at best a surrogate measure of WLBD risk and at worst a distraction from better predictors of injury, such as combined or complex dynamic movements, modeled tissue strain, and strain/load rates. Similarly, Hoozemans *et al.* (2004) described "net moment" as an output which gives no information about the strain on various tissues and is of limited value because tissue strain is more likely related to injury. Even though the compression forces may not be the major cause for the precipitation of WLBD, using compression as an indicator to assess the risk of WLBD is a useful approach (Norman and McGill, 1993). Compression not only demonstrates loading of the intervertebral disks, but is related to increased loading of tissues such as muscles, tendons, and ligaments and may be valuable as a surrogate measure of these factors. McGill (1997) stated that "combining biomechanical modeling techniques to obtain tissue loads with studies of tissue mechanics and structural architecture is a powerful approach for analyzing injury mechanisms, assessing the injury risk and preparing injury avoidance strategies". Table 5-2 presents biomechanical models described in the literature classified by type (static, quasi-dynamic, and dynamic, respectively). Only the models identified by our limited literature search were reviewed. We believe that the most commonly-used models were included, but it is likely at least some models were missed. Many models were knowingly excluded; exclusion does not necessarily represent low model quality but rather impracticability for workplace setting use; excluded models may be both important and conceptually relevant.

Table 5-2. Biomechanical models to estimate work-related low back loading in occupational settings identified in the literature.

Reference	Model type	Main Inputs	Main Outputs	Strength	Limitation
Schultz *et al.* (1981)	Static; 3-D; optimization	Hand forces; anthropometry; positions of joints	3 net forces and moments: antero-posterior and lateral shear, and compression	Authors suggest model can be applied to low-acceleration movements	Assumes single equivalent muscle, with longer moment arms estimated to produce more moment; does not consider co-activation
Schultz and Andersson (1982)	Static; Link-segment	Lift parameters; anthropometry	Muscle tension; L3 compression	Muscle activity closely reflects calculated net flexion moment; iterative calculations; analyzes vertical and horizontal hand forces	Approximates extensor muscles using single equivalent muscle; assessment of symmetrical sagittal movements only
Andersson *et al.* (1985)	Static; Link-segment	Percentage of maximum range of motion for spine, sacrum and leg; anthropometry; hand forces	Estimates L5/S1 orientation, IAP, disc, ligament and muscle contributions to extensor moment	Continuous data allows time history of task and both static and dynamic calculations; allows comparison of static and dynamic estimates of loading	Symmetric sagittal lifting movements only; approximates back extensor muscles using single equivalent muscle
Hoozemans *et al.* (2004)	Static; Pushing and pulling	Direction of forces (push or pull); one or two handed; cart weight; handle height	Compressive and shear forces in the lumbar spine both for initial force and sustained force	Simple; easily measured inputs; practical for field use even for large numbers of assessments; developed using an EMG-assisted model	Limited to pushing/pulling; only applicable to hip and shoulder handle heights
Chaffin and Anderson (1991)	Static; 2 or 3D; direct dynamics	Positions of 15 joints (picture or video); anthropometry; load size and position; hand forces and their direction	Moments; compression and shear at L4/L5 and L5/S1; muscle strength requirements to maintain major joint postures	Relatively simple to use; commercially available software application; anthropometrical and psychophysical database included allowing comparisons between strength capabilities and % of the population capable of maintaining the posture	Underestimate loads during highly dynamic activities; limited for infrequent tasks and/or peak loads

Table 5-2. Biomechanical models to estimate work-related low back loading in occupational settings identified in the literature. (cont...)

Reference	Model type	Main Inputs	Main Outputs	Strength	Limitation
Sullivan *et al.* (2002)	Static; 2D; direct dynamics	Positions of 7 joints (video)	L4/L5 compression, shear, and moment	Simple to use, small number of inputs required	Underestimate loads during asymmetrical and/or highly dynamic activities; limited for infrequent tasks and/or peak loads
Newell and Kumar (2005)	Static; 2D; direct dynamics	Postures of 3 segments (trunk, neck, shoulder) (video recordings)	Compression and shear forces on the lumbo-sacral and cervico-thoracic joints	Allow for cumulative load estimates in stationary tasks; applicable for sitting jobs	May underestimate loads during asymmetrical and/or highly dynamic activities; limited for stationary, or infrequent tasks, and/or peak static loads; susceptible to joint marking mistakes
Norman *et al.* (1994)	Quasi-Dynamic; 2 or 3D; direct dynamics; optimization	Positions of 20 joints; hand loads (force transducer)	Forces, moments, compression and shear at L4/L5	Available commercially; tested against a complex model outputs	Does not account for co-contraction
McGill and Norman (1985)	Static, quasi-dynamic, and dynamic	Positions of joints; anthropometry; hand forces	Moments about L4/L5	Allows comparison of static and dynamic estimates of loading and the use of most appropriate mode	Sagittal only
Morlock *et al.* (2000)	Quasi-Dynamic; 3D; EMG-assisted; inverse dynamics	Postures of 8 segments (electrogoniometers); bilateral EMG (rectus abdominis, external oblique, latissimus dorsi, and erector spinae); ground reaction forces (force insoles)	Moments and torques; compression and shear forces	Detailed loading information; continuous kinematic information	Underestimate loads during highly dynamic activities; expensive equipment required to obtain input data; more complex, less practical; insensitive to forces applied below thoracic spine; not applicable for sitting tasks

Table 5-2. Biomechanical models to estimate work-related low back loading in occupational settings identified in the literature. (cont...)

Reference	Model type	Main Inputs	Main Outputs	Strength	Limitation
Andres and Chaffin (1991)	Dynamic; 2D	Motion of joints	Compression at L4/L5; floor coefficient of friction required	Sagittal asymmetry permitted, tested against inverse dynamics EMG-assisted model; includes co-contraction effects	Sagittal plane tasks only, over-predicts ground force during pushing
Potvin (1990)	Dynamic; EMG-assisted	EMG from the erector spinae; calibration procedure relates to linear prediction of spinal loads based on estimates from a static biomechanical model	Spinal compression	Portable EMG systems allow long-term, continuous measurement and cumulative load estimates	Assumes that relationship between static contractions and EMG applies to dynamic contractions
Fathallah *et al.* (1999)	Dynamic; 3D	Anthropometry, trunk motion (lumbar motion monitor), measured moments	Continuous compression, antero-posterior and lateral shear	3-D continuous data; no EMG required; intended for eventual use in industrial applications; tested against EMG-assisted model	Reliability of the shear estimates form the simplest kinematic model is modest; asymmetric motions predictions are less reliable than symmetric estimates; shear underestimated in general; tested in limited number of simulated and industrial conditions
Nussbaum *et al.* (1999)	Dynamic; 3D; link-segment; direct dynamics; optimization	Motions of seven segments; external force (force transducer)	L3/L4 force, moments, compression and shear forces	Allows arriving at a solution with the inclusion of multiple muscles in the model	Optimization process based on minimum muscle contraction to stabilize the system (may not reflect actual behavior)
Mientjes *et al.* (1999)	Dynamic; EMG-assisted	EMG from the erector spinae	Compression forces	Allows estimation of cumulative loads	Reduced accuracy for non-sagittal and asymmetric tasks
Skotte (2001)	Dynamic; 3D; inverse dynamics	Motions of 7 body segments (12 markers), ground reaction force (force platforms), contact points reaction force (force transducers on bed frame for example)	L4/L5 net moment, compression and shear forces	Includes the contacts between the lower limbs and furniture (bed); includes cross-sectional area data of 14 muscles (database)	Does not account for co-contraction; complex; requires instrumentation of furniture or machinery for contact reaction forces

5.3.3. Static Biomechanical Models

Schultz and Andersson (1981) developed a static, link-segment model which uses information on joint positions and hand forces to compute three net forces and three net moments about the L3 vertebra. Because there are six outputs from this model, the calculations require enough measured or set values as inputs to avoid indeterminacy. Spinal forces are computed from moments using a single equivalent extensor muscle approximating the erector spinae acting with a set moment arm of 5 cm (Morris *et al.*, 1961).

The Static Strength Prediction Program (3D SSPP) is a software package including a static, two or three-dimensional biomechanical model of the spine (Bean *et al.*, 1988; Chaffin and Andersson, 1991). The model uses a double-linear optimization approach to provide estimates of moments, compression and shear forces at L4/L5 and lumbosacral joints, and estimates of the muscle strength requirements to maintain the system in static equilibrium. The calculation inputs include kinematic data from thirteen joints (ankles, knees, hips, trunk, shoulders, elbows, and wrists), three anthropometric characteristics (height, weight and gender), and six aspects of the external load (weight, position, vectors magnitudes and directions). The program estimates the percentage of the population capable of performing the exertion by comparing the muscle strength requirements with a database of strength capability data for the USA adult population (Chaffin and Andersson, 1991).

Waters *et al.* (1998) stated that the main limitations of the 3D SSPP are that it is "not applicable to repetitive activities", it is "not applicable to highly dynamic activities", and that it is "difficult to obtain input data". However, repetitive tasks may be analyzed using a frame analysis procedure combined with adequate time and frequency multipliers. In addition, all biomechanical models require relatively complex input data and the 3D SSPP input requirements are among the simplest. As for strengths, Waters *et al.* (1998) stated that the program "provides detailed estimates of mechanical forces on the body", and it "can identify specific structures exposed to high risk"; the 3D SSPP provides estimates of compression, shear, percent of the population capable, and percentage of ligament strain associated with the task.

The 3D SSPP has been used to estimate strain in the lumbosacral fascia and compressive forces at the lumbosacral joint in chiropractors using data collected with the lumbar motion monitor, video-recordings, and photographs (Lorme and Naqvi, 2003). The 3D SSPP has been compared to many other exposure assessment methods in occupational field studies. Waters *et al.* (1998) assessed warehousing jobs using the 3D SSPP, the lumbar motion monitor and associated WLBD risk assessment model, a psychophysical manual lifting assessment, and the National Institute for Occupational Safety and Health (NIOSH) lifting equation. All methods identified the warehousing jobs as having high risk for WLBD. However, there were differences between the estimates of each method. The authors recommended that the 3D SSPP should be used only for assessing infrequent tasks because it was not originally designed to consider the cumulative effects or the multi-task characteristics of manual materials handling. Similarly, the opposing results may be due to the different characteristics of the jobs analyzed in the studies (metal fabrication vs. rescue tasks).

Lavender *et al.* (1999) stated that the 3D SSPP results may underestimate the risk of cumulative WLBD because the model was originally proposed to assess peak loads resulting in risk of overexertion injuries. The authors argue that it would be extremely time-intensive to calculate the load for extended periods using 3D SSPP, while other methods partially account for the cumulative risk of injury. Mirka *et al.* (2000) evaluated the biomechanical stress of construction workers using the revised NIOSH lifting equation, the 3D SSPP, and the lumbar motion monitor risk assessment model and found different results depending on the method used. The differences may come from comparing the results for specific task components as opposed to the overall job demands. In addition, these are entirely different models that use distinct criteria and, consequently, different results.

Lavender *et al.* (1999) compared estimates of WLBD risk based on the NIOSH 1993 lifting equation, the 3D SSPP, the lumbar motion monitor risk assessment model, and two variations of the United Auto Workers – General Motors Ergonomic Risk Factor Checklist in a metal fabrication company. The 3D SSPP L4/L5 compression force outputs were used to classify the job tasks as low risk (< 3433 N), medium risk (from 3433 N to 6377 N), or high risk (> 6377 N). The highest correlation was between the 3D SSPP and lumbar motion monitor classifications (r = 0.39), but the correlation was still low. However, in a subsequent study, Lavender *et al.* (2000) assessed the biomechanical load of frequent rescue tasks by firefighters using the 3D SSPP and the lumbar motion monitor risk assessment model and found good correlation between the estimated L4/L5 compressive forces and the lumbar motion monitor predicted probability of high WLBD risk (r = 0.78).

Silvia *et al.* (2002) compared the low back compression estimates for patient transfers using two mechanical lifts and manual transfers. The compression estimates were calculated using the 3D SSPP and a dynamic, EMG-assisted biomechanical model. The input kinematic data for the 3D SSPP was collected by video-recording six subjects performing the transfers. The video was reviewed "to identify the most stressful postures associated with each task". The EMG-assisted model used the EMG of the erector spinae muscles to estimate the low back compressive forces using linear regression. The regression equation was calculated based on two calibration trials holding 8.8 lbs and 14.3 lbs loads in neutral and 45° of flexion. The compression forces for each of these calibration trials were estimated using the 3D SSPP. The regression equation explained between 63% and 99% of the variability (average r^2 = 0.863). The peak 0.33 seconds of compression estimated by the EMG-assisted model were identified for each transfer. In general, the authors found that the low back load was lower when using the transfer devices and found that the use of one of the lifts resulted in reduced low back loads. There were differences between the compression forces estimated by the 3D SSPP and by the EMG-assisted model. The latter had bigger standard deviations and tended to estimate higher compressive forces. This may be due to the fact that the continuous EMG recording used may have indirectly captured the increased loading due to the dynamic aspects of the transfers which are neglected by the 3D SSPP.

Hoozemans *et al.* (2004) developed a sagittal, static biomechanical model to determine compressive and shear forces in the low back during pushing and pulling tasks in the workplace. Hand forces from transducers and joint positions from light emitting diode markers were used to develop a model with easily acquirable inputs: handle height, cart weight, one or two-handed effort, and push/pull direction. Practical inputs result in a practical, simple to use model for large numbers of assessments. Unfortunately, this push/pull model is only applicable for hip or shoulder-level handle heights.

Anderson *et al.* (1985) developed a static, link-segment biomechanical model to predict L5/S1 loading during sagittal, static lifting tasks. The model incorporated passive tissue effect in moment generation. The forces resulting from the posture and load in the hands were balanced by five ligaments, intra-abdominal pressure, disc resistance, and a single equivalent extensor muscle representing the erector spinae with a 5 cm moment arm. An iterative computation is used to account for the interaction between the tissues; this is necessary because disc height affects ligament strain, and compression affects disc height. First, compression is estimated considering only the torques from body segment mass and the load in the hands. Then, this initial compressive load is used to estimate disc height, which in turn is used to estimate ligament length. The ligament length is estimated from body position and the strain is calculated by comparing resulting and resting ligament length. If total compression is not within 1 N of the estimates including disc and ligament forces, another compression value is used; this process is reiterated until the values are within 1 N. This model is currently not available commercially which may restrict its application.

5.3.4. Quasi-Dynamic Models

Morlock *et al.* (2000) proposed a quasi-dynamic, three-dimensional, inverse dynamics, EMG-assisted biomechanical model to assess lumbosacral workload and tested the model in twelve nurses during regular work. The data was collected continuously during work for four hours. The model includes eight rigid bodies from feet to lumbar spine as well as bilateral actions of the rectus abdominis, external oblique, latissimus dorsi, and erector spinae muscles. It incorporates kinematic input data from electrogoniometers and a torsiometer for joint motion and kinetic input data on ground reaction forces from force distribution insoles. The external forces were represented by one-dimensional ground reaction forces which may cause overestimation of shear forces. The model does not take into account external forces acting below the thoracic spine which may underestimate the load if the body weight is partially supported (e.g.

underestimating loads during sitting), and some joint motions are not included (such as hip rotation and inversion/eversion).

Norman *et al.* (1994) developed a sagittal, quasi-dynamic link-segment, optimization model to be used in the workplace. The model is available as a software program called WATBAK. The model allows the estimation of loads (forces and moments) on the wrists, elbows, shoulders, C7, and L4/L5. It calculates the compression and shear forces at L4/L5 from the muscle moment using a 6 cm moment arm for the extensors. The model was classified as quasi-dynamic because it includes the inertial forces of the external loads but it does not account for the acceleration of the body segments. Anthropometric data can either be entered by the investigator or selected from a database of male and female percentiles; postures can be entered as x, y coordinates or by positioning a manikin on the screen.

Cole *et al.* (2004) evaluated the lifting component of a work capacity assessment test using the WATBAK program and EMG for the erector spinae at L4 and rectus abdominis. Sagittal video recordings and kinematic data from the Peak Motus 2000 system were collected from six subjects performing five lifting tasks. Although the model was not compared with other measures, the authors state that the actual loads may be underestimated because the model neglects the inertial effects on the body segments, and the model does not consider abdominal co-contraction. Kerr *et al.* (2001) compared the lumbar loads during occupational activities performed by 137 workers with WLBD (cases) with the loads on 179 workers without WLBD (controls) using the WATBAK model. They identified peak lumbar shear force (OR = 1.7, 95% confidence interval = 1.02-2.86), peak load handled (OR = 1.9, 95% confidence interval = 1.21-3.10), and cumulative compression (OR = 2.0, 95% confidence interval = 1.22-3.59) as biomechanical risk factors for WLBD, demonstrating the model's ability to identify risk factors to WLBD.

5.3.5. Dynamic Models

Although static models are often applied to work tasks, the majority of occupational tasks are dynamic. Dynamic biomechanical models account for the inertial forces due to acceleration and deceleration of the external loads and body segments.

Dynamic models suitable for industrial applications may incorporate data from EMG, force transducers, joint movement position, or a combination.

Optimization models: Andres and Chaffin (1991) developed a dynamic biomechanical model to predict both compressive forces on the spine and friction coefficients needed to avoid slips and falls. Although the model was designed for sagittal plane tasks only, it accommodates the asymmetry that comes with walking (e.g., one leg in front of the other), and it provides estimates for a one-legged or two-legged stance. The model uses force plate and kinematic data from joint markers to predict erector spinae and rectus abdominis torques and estimate compressive loads about L5/S1. The model assumes that all muscle forces are parallel to the spine and produce compression only, but the moment arm adjusts according to hip angle and it incorporates intra-abdominal pressure effects on the extensor moment. Predictions for one-legged stance were better than two-legged stance. Predicted rectus abdominis muscle torques correlated well with EMG, but erector spinae muscle torques were underestimated. The correlation between predicted and measured ground reaction forces was moderate ($r = 0.67$), and there was more overestimation of ground reaction force during pushing than during pulling.

Skotte *et al.* (2002) evaluated the L4/L5 net moment, compression and shear forces during nine different patient handling tasks including turning, lifting, and repositioning by ten female health care workers using a dynamic, three-dimensional biomechanical model. The model used inverse dynamics and included seven body segments, and the cross-sectional area data of fourteen muscles (Skotte, 2001). The input information for the model was gathered using a five-camera video recording system, force platforms, and force transducers on the bed. The authors did not directly assess the biomechanical model used, but they mentioned as a limitation that it does not include the effect of muscle co-contraction that may increase the compressive forces on the lumbar spine. On the other hand, the authors measured the reaction forces resulting from the contact of the worker with the bedside during the task. These reaction forces were included in the biomechanical model calculations and may have increased the accuracy of the results. By using a three-dimensional dynamic model they accounted for the asymmetrical loading of the spine and for the effect of rapid motions. The five-camera system may restrict its application in some industrial settings.

EMG-assisted models: Most frequently occupational work is dynamic, involving the co-contraction of multiple muscles and concomitant intra-abdominal pressure. The torques, moments, and resulting forces change throughout the motion. It is challenging to incorporate all these factors into a biomechanical model that can be used in the workplace setting. EMG of agonist, antagonist, and synergist muscles may provide an avenue to analyze twisting, lateral bending, and differential loading. EMG may also permit continuous measurement and estimation of instantaneous loads after calibration, as well as the estimation of cumulative loads.

Some studies on EMG-assisted biomechanical models have employed data collection techniques which required measurement of dynamic lifting while the pelvis was restrained within an apparatus, limiting the realism of the movements under study (e.g. Granata and Marras, 1995). Unconstrained free dynamic lifting was studied by Fatahllah (1997, 1998) using a force plate combined with a pelvic orientation monitor to determine the three-dimensional orientation of the pelvis. In the conditions tested, the average percent error in estimating the actual applied moment was about 4%. This method accounts for complex three-dimensional movements, but as the worker must stay on the force platform, this method is useful only for single lifts or stationary work. However, force insoles may be used to allow more freedom to the workers to move around.

Three dimensional, EMG-assisted models incorporating joint motion and external force data have been used as a reference or "gold standard" in the development of simpler models to estimate dynamic loading (e.g. Fathallah *et al.*, 1999). However, Marras and Sommerich (1991) pointed out the need for a better understanding of the EMG-force relationship to enhance EMG-assisted models. In addition, Marras and Granata (1995) stated that other drawbacks of EMG-assisted models are that they are "more complex and cumbersome than many of the commonly used ergonomic models", and EMGs "are not always feasible to obtain in the workplace." Nussbaum *et al.* (1999) compared a dynamic, three-dimensional, seven link-segment, optimization model using direct dynamics with a dynamic, EMG-assisted biomechanical model (Nussbaum and Chaffin, 1998). The models were tested in a lab setting to estimate L3/L4 moments, compression and shear forces, and muscle antagonism during manual material handling using lifting devices. For the optimization model, kinematic input data for the upper and lower arms, thoracic and lumbar spine, and pelvis was collected using a four-camera motion analysis system. The kinetic input data was collected using a force transducer between the lifting aid and load. Continuous EMG was recorded bilaterally for the erector spinae (lumbar and thoracic), latissimus dorsi, internal and external oblique, and rectus abdominis muscles for the EMG-assisted model. The association between the estimates of lumbar moments from the EMG-assisted model and from the optimization model was assessed using linear regression and the root mean square differences ($RMS_{dif.}$) between the models were calculated for the sagittal ($r^2 = 0.764$, $RMS_{dif.} = 14.14$ Nm), frontal, and transversal planes. Although results for frontal and transverse planes were not reported, the authors stated that the correspondence between the moments were lower in these planes and in asymmetrical tasks. The differences between the models were attributed to the effects of co-contraction of the antagonist muscles which were accounted by the EMG model but ignored by the optimization model; optimizations models assume a minimum muscle recruitment intensity (muscle force divided by muscles' cross-sectional area) to minimize the spine loading. The difference between the models was used as an index of "muscular antagonism" which was higher during the use of a hoist manipulator in comparison to an arm manipulator and to manual transfers. The general (all planes) root mean square difference was approximately 10 Nm. The strength capabilities were based on static data. The authors do recognize that the results may have overestimated the strength capabilities because it has been shown that dynamic strength is lower than static (Kumar *et al.*, 1988).

Potvin *et al.* (1996) developed a dynamic EMG-assisted biomechanical model for workplace assessments. The authors compared two dynamic EMG-assisted models, one including kinematic data and one only using EMG. The EMG-only model predictions of peak loads during eccentric and concentric lifting were respectively 9% (SD = 4) and 26% SD = 12% lower than the predictions from the EMG with kinematic data model. Mientjes *et al.* (1999) also developed an "EMG-only" method to estimate spinal loading during occupational activities. This model uses a linear transformation of EMG into compression forces based on the estimates from a two-dimensional biomechanical model (2D WATBAK). The regression equation used in the transformation was calculated during a calibration test at 60° of sagittal trunk flexion holding an external load of 22 kg for 5 s. The amplitude probability distribution function is used to examine the proportion of time spent at different loading levels. Lab tests of simulated lifting were used to compare estimates from this EMG-assisted model with those from a three-dimensional biomechanical model (3D WATBAK). The input data for the 3D WATBACK was collected with a hand-force transducer and a motion analysis system (Peak Performance) using two synchronized video cameras to record the bilateral positions of the following landmarks or joints: ear canals, C7/T1, shoulders, elbows, wrists, hands, L4/L5, hips, knees, ankles, and fifth metatarsals. The average difference in spinal compression was 14.9% for all tasks and 30.7% for twisting tasks. Similarly, the percentage of time spent in various loading levels was, on average, within 6.5%; this average difference increased to 13.4% during twisting. Co-contraction is known to occur in many tasks and particularly in twisting. However, this model is based only on the activity of the erector spinae muscles (mainly a trunk extensor) at T9 level and did not include the rotational muscles such as the external oblique abdominals. The strength of this EMG-assisted model is its ability to estimate the proportion of time spent at a given compression level over varied tasks over extended periods of time. On the other hand, the accuracy of the instantaneous compression estimates, especially during twisting tasks, is limited. Calibrated or "compression normalized" EMG-assisted models allow for long term, continuous measurement using portable EMG systems and have been used successfully to measure peak and cumulative load in residential care aides (Village, 2005) and in sheep shearers (Marshall, 2005).

5.4. Discussion of Chapter 5

5.4.1. Validity and reliability of biomechanical models

Imprecision in spinal load estimates is inescapable. As pointed out by Keyserling (2000), assumptions are necessary to simplify a biomechanical model or the solutions become indeterminate. In general, the higher the complexity associated with a biomechanical model, the higher its resolution. However, complex models may be restricted to laboratory settings. An ideal model meets the objectives of a study or job assessment with a balance between accuracy and applicability to actual working situations.

The criteria used to assess the reliability and validity of biomechanical models' estimates requires consideration. Concurrent validity requires comparing results from simultaneous testing of the same phenomenon using a valid, gold-standard reference, while parallel reliability requires the same simultaneous testing comparisons using a consistent (reliable) reference (Gadotti *et al.*, 2006). A measure can be reliable but not valid, but a measure cannot be valid if it is not reliable. Biomechanical models should have high content validity in their anatomical and physiological aspects. However, assumptions and simplifications are inevitable and are present in all models. In vitro tests of tissue mechanical properties and injury tolerance of spinal segments preparations have been used to determine cut-points where the risk of injury increases significantly (e.g. Waters *et al.*, 1993; Vieira and Kumar, 2006). The main limitation of this strategy is that the in vitro musculoskeletal tissue behavior may differ from in-vivo. According to Keyserling (2000) "cadaver studies have been questioned because it is not known whether the behavior of cadaver tissue is similar to that of living tissue".

Testing the concurrent validity of spinal compression estimates from biomechanical models may involve using a pressure gauge needle inserted into the intervertebral disc (Nachemson, 1966; Nachemson and Elfstron, 1970). It was demonstrated that the hydrostatic pressure within the intervertebral discs increased linearly with increased compression forces acting on the spine (Berkson *et al.*, 1979). Schultz *et al.* (1982) also compared in-vivo measurements of intradiscal (L3) pressure to biomechanical model estimates of compressive load; the measured pressures and predicted values were well-correlated. Based on these results, the concurrent validity of force estimates of biomechanical models applicable to workplace settings may now be sought using comprehensive lab-based biomechanical models, such as the EMG-assisted stochastic model described by Mrika and Marras (1993). Marras (2000) stated that "3D, EMG-driven models are the most accurate biomechanical models available at the moment to estimate low back loading". However, Schultz *et al.* (1982) found less EMG activity per unit of contraction force when flexed than when upright, indicating some caution may be needed when interpreting EMG activities quantitatively, as the assumption of linearity in tension-activity relationship may be only partially valid over wide ranges in significant muscle length changes. In addition, the EMG assisted models substantially overestimate compression and shear forces when acceleration of movement is present.

The complexity of some EMG-driven models may limit their applicability, their value is in incorporating different factors such as muscle activity/coactivity (Baten *et al.*, 1996; Dolan and Adams, 1993; Kingma *et al.*, 1996; Marras and Granata, 1995 and 1997); trunk motion characteristics (Marras *et al.*, 1991; McGill, 1990); spinal and pelvic position (Marras and Granata, 1995); ground reaction forces (Fathallah *et al.*, 1998; Kingma *et al.*, 1996); hand forces (Raschke and Chaffin, 1996); intra-abdominal pressure (Marras and Reilly, 1988), and population-based data on typical variability of loading (Reilly and Marras, 1989). The data collection schemes of complex dynamic models are not easily applicable to workplace settings, but the understanding gained from laboratory-based experiments can be applied to work providing insight into the nature of loading and the anatomical, postural, and task-based determinants which might be addressed. Unfortunately, without a "gold standard" biomechanical model feasible for workplace validation, even the most complex spinal load models used in occupational settings only provide estimates of the spinal loads rather than provide precise calculations of the absolute force levels (Waters 1993).

The lack of a "gold standard" limits evaluation of the models' concurrent validity, although models can be compared to one another for relative differences (parallel reliability). In the words of Wells (1997) "none of the spine models that have been presented to date in the literature have been directly validated by comparison of model estimates of muscle force, spinal compression or shear with direct measures of these variables in the same units of measurement." Predictive validity of models' in relation to health outcomes further illustrates difficulty in selecting a "gold standard." Comparing exposures as predicted by a biomechanical model to risk category classification, such as the NIOSH lifting index or the LMM risk model, carries an inherent circularity in the layers of assumptions; if the initial classification into risk categories is made based on biomechanical models or measures, the relationship to a new biomechanical assessment is likely to be strong. To validate methods with respect to health effects, the exposure levels and prevalence rates, or preferably to prospective incidence of WLBD need to be compared. Some of the challenges are the multifactorial nature of WLBD, the confounding effects of non-occupational exposures, recreational activities, psychosocial issues and personal habits such as smoking and exercise (Vieira *et al.*, 2006). Complete validation of biomechanical models has not been achieved. Generally, the biomechanical models have demonstrated low predictive validity. Future studies and developments are required to overcome this significant problem.

In relation to intra and inter-subject reliability, it is important to recognize that human motor functioning is inherently variable. Even when the same occupational task is executed by different subjects, variations in performance and conditions can have considerable effect. There is between-subject variation in recruitment patterns as well as within-subject variation over repeated trials of the same task (Marras and Granata, 1997; Marras *et al.*, 1999; van Dieen *et al.*, 2001). In a laboratory experiment of simulated tasks, Granata and Marras (1999) found that repetitions of the same task performed in the same set of conditions "produces a wide range of loads", with more inter-repetition variability when tasks were asymmetric. Even when the mean compression estimate of repeated lifting trials was below the NIOSH action limit (3400 N), 20% of repetitions exceeded this value. The authors stated that disregarding variability underestimates the "relative number of repeated exertions that might exceed" the limit. One can imagine that under self-paced workplace conditions with free dynamic lifts, a single assessment of physical load (however accurate for that lift) might not represent the typical range of exposures.

Sullivan *et al.* (2002) evaluated the intra- and inter-tester reliability of the L4/L5 compression, shear, and moment estimates using a static, two-dimensional biomechanical model. The hand, wrist, elbow, shoulder, ear canal, C7/T1, and L4/L5 joints were marked on videos from a sagittal lifting task performed by one subject wearing tight, dark clothes. The videos were marked by ten different testers and by the same testers at five different times over three different days to provide the kinematic information required by the biomechanical model. The authors found no significant intra- or inter-tester differences using ANOVA and good agreements using intra-class correlation for the testers, days, and trials. Compression and moment estimates presented the highest reliability while reaction shear presented the lowest reliability. The study was performed in a controlled laboratory setting; the reliability could have been lower if the testers were marking videos from an actual industrial task with asymmetrical components and regular clothing. However, the reliability of joint marking while wearing work clothing has been investigated; Newell and Kumar (2005) compared the angles marked on two subjects wearing only a brief with the angles marked when they were wearing regular clothing. No significant differences were found.

5.4.2. Peak loads, cumulative loads, and continuous measurement

The time history of physical loading is important to determine not only the magnitude of loading but also the duration, repetition, and work-rest pattern (Wickstrom *et al.*, 1996). Therefore, those methods which allow continuous evaluation of spinal loading are preferred when a global picture of exposure and its relationship to injury are desired. Continuous joint position measurement allows static models to estimate cumulative loads as well as instantaneous peak loads (Kumar, 1988; Newell and Kumar, 2005), even though there is some level of underestimation due to the unaccounted inertial forces.

Peak and cumulative loads are both important, independent risk factors for WLBD (Kumar, 1990; Norman *et al.*, 1998). Daynard *et al.* (2001) showed that, despite reducing the peak load on the nurses' lumbar spines, the use of some patient lifting devices may increase the total cumulative load because of the longer time required to perform the patient transfers. Cumulative compression and shear forces were shown to be higher in institutional aids with low back pain than in those without pain (Kumar, 1990). When continuous measurement and on-line estimation is not available or possible, more labor-intensive serial estimations of loading using sequential frame analysis from video or still pictures can be done.

5.4.3. Matching the model to the job tasks analyzed (static vs. dynamic)

Spinal loading varies with the rate of lifting, as tasks become more dynamic there are higher accelerations (Chen, 2000). Static models are designed to estimate spinal loads during stationary postures or holding efforts. Although they tend to be the simplest and most practical in terms of inputs and calculations, their main drawback is that the majority of occupational tasks are dynamic. Spinal loads are underestimated when static models are applied to dynamic tasks because these models do not account for the inertial forces (Wickstrom *et al.*, 1996). These models assume that the system (lumbar spine for example) is in static equilibrium and they do not take into consideration the effect of the acceleration moments.

Comparisons of static and dynamic models report considerable differences: static model estimates of peak lumbosacral compression being 33 to 60% lower than dynamic models" results (Leskinen, 1985). Dynamic model estimates have been reported to be 1.5 times higher (Jager *et al.*, 1989), 19-52% higher (McGill, 1985), and up to 200% higher (Waters *et al.*, 1993) than static model estimates for the same task. The differences are more pronounced when the task is more dynamic. Even during slow controlled tasks, Nussbaum *et al.* (1999) found estimates of the percentage of the population who could perform a dynamic task to be 5 to 10% lower when a dynamic model was used compared to results from a static model, demonstrating the underestimation of loads due to body segments motion when static models are used.

Quasi-dynamic models approximate dynamic loads more closely than static models, but quasi-dynamic models only consider the inertial forces of the external loads, not body segment inertias. Skotte *et al.* (2002) found that the differences between dynamic and quasi-dynamic model estimations were less than 5% in 90% of the cases, but dynamic estimates of torque were 17% higher with sudden, rapid pulls. Detailed loading estimates for highly dynamic tasks require a model complex enough to account for acceleration. However, when a more approximate estimate will suffice, static or quasi-dynamic models may provide enough information.

As described by Schultz and Andersson (1981) "…the compression load on the lumbar spine, for example, might be near zero in quiet lying, 400 N in quiet standing, and 4000 N in a strenuous exertion. Given the range of loads experienced is so large, even a rough estimate of the loads generated by an activity will usually suffice for the solution of practical problems." The authors also state that "many activities involving body motion can be analyzed without significant error as if they were static" and that more complex dynamic models are required "only when motion involves significant linear or angular accelerations". The utility of the static biomechanical models is their ability to estimate and categorize the broad range of human efforts rather than to distinguish small changes in loading.

5.4.4. Anatomical simplifications and plane of motion assumptions

Most link segment models make assumptions to simplify anatomy, but these assumptions can limit the accuracy of the estimates. For example, most muscles do not have straight, vertical lines of actions, but work around a pulley of bones or other muscles (McGill and Norman, 1986); the erector muscles do not attach adjacent vertebrae as in many simplified models; single equivalent extensor muscles do not reflect the action of muscles over broad areas, and a single equivalent moment arm of 5 cm is not anatomically accurate (Kumar, 1988). The latissimus dorsi are neglected in many models, although they have the longest moment arm of all the extensor muscles (McGill and Norman, 1986).

On comparing the outputs of different biomechanical models, Skotte *et al.* (2002) found differences in shear forces and explained that these values vary according to the anatomical representation used (especially for the erector spinae line of action angle) in the biomechanical model. Single equivalent extensor muscles are generally limited to sagittal plane analyses and cannot account for forces which are not parallel to the spine. However, axial twisting and shear forces have been suggested as risk factors for the spine, and are therefore worthy modeled outputs.

Estimating lumbar loads during multi-planar tasks is complicated. No single muscle produces trunk axial torsion for example, and muscles change lines of action during twisting. Therefore, planar estimations are not ideal for multi-planar tasks. McGill (1991) points out that "out-of-plane postures contribute a significant amount of daily activity", but most static biomechanical analyses are planer. Skotte *et al.* (2002) showed that for the same task the compression estimate using a 3D model was 3330 N while the compression was 2290 N for a sagittal model. EMG-assisted biomechanical models may help to avoid some of the assumptions which constrain movement to the sagittal plane and neglect the actions of some muscles, thereby allowing the study of complex and compound movements limiting the amount of body motion data (kinematic data) required.

Forces and moments can be estimated using optimization algorithms to minimize the compression forces on the spine or the strength requirements of the back extensor muscles (Schultz and Andersson, 1981). However, without other constraints, the "optimum" solution will load the muscles with more advantageous moment arms preferentially. This tends to underestimate the amount of co-contraction of synergists and does not take into consideration any co-contraction of the antagonist muscles. Schultz and Andersson (1981) noted that in sub-maximal efforts the true compressions can vary widely between repetitions and from the "optimized" estimate, depending on differences in muscle recruitment. The success of the linear program method depends on the quality of the underlying anatomical model. Maras and Somerich (1991b) noted that "optimization models usually cannot account for coactivity because the quantity of model solutions is limited by the number of functional constraints."

5.4.5. Presence and effects of co-contraction

According to Kingma *et al.* (2001) "the most widely used and extensively validated non-invasive method of quantifying spinal loading is the linked segment model". However, this method (when used in isolation) does not take into consideration co-contraction of the antagonist and synergist muscles resulting in underestimation of the spinal loads. Co-contraction during trunk extension stabilizes the spine, but it also causes increases in muscle fatigue and spinal compression up to 21% (Gardner-Morse and Stokes, 1998). Similarly, according to Thelen *et al.* (1995), the co-contraction during static trunk extension exertion increases the lumbar spine compression forces between 16% and 19%, and abdominal muscle co-contraction during dynamic extension has been shown to increase shear forces by 70% (Granata and Marras, 1995).

Co-contraction effects can not be accounted unless the moments or forces by antagonistic muscles (such as the rectus abdominis during extension) are measured or assumed. Schultz *et al.* (1982) found a high correlation ($r = 0.987$) between predicted contraction forces using a two-dimensional model and EMG from lumbar trunk and abdominal muscles in twelve tasks. Antagonistic muscle activity was observed only with horizontal (push/pull) forces.

Schultz *et al.* (1982) reported "more than 4/5 of the spinal compression was due to the muscle contraction forces in the most stressful tasks", with limited contribution by passive elements. Including antagonist co-contraction effects in biomechanical models increase the accuracy of the low back load estimates. EMG facilitates the collection of muscular co-activation data. Modeling multiple muscles require the incorporation of anatomical data on muscle location, cross-sectional area, and line of action, but in optimization models this level of complexity would have the number of unknown muscle forces exceeding the number of equations and render the solution indeterminate. In setting some of these muscle forces to zero, the synergistic and antagonistic co-contractions are ignored and so are any voluntary recruitment strategies which account for differences between individuals.

EMG-assisted models can help to address the indeterminacy by attributing forces to different muscles, and by accounting for individual differences in task execution and muscle recruitment patterns. Marras and Granata (1995) measured EMG of five trunk muscles and compared biomechanical model predictions including different combinations of muscle inputs. The level of co-contraction depended on the weight lifted and trunk velocity; co-contraction significantly altered the estimated load. Optimization strategies not considering the effects of co-contraction underestimate the spinal load. Static models using such optimization strategies present a dual threat to precision under dynamic conditions: underestimation of spinal loading due to co-contraction and due to inertial forces.

5.4.6. Effects of contact forces between subject and furniture

The estimated moments on the low back may be inaccurate when using inverse dynamics models if the external forces applied to the lower limbs are not included in the calculations (Morlock *et al.*, 2000). Many jobs involve the contact of different body parts with the furniture or machinery. The vast majority of the biomechanical models do not take into consideration these external forces. Thus, the calculations may have different degrees of error depending on the amount, magnitude, and location of contact.

Skotte (2001) evaluated the effect of bedside reaction forces on L4/L5 moments on nurses during patient handling. The net moments were estimated using an inverse dynamics, dynamic, three-dimensional biomechanical model using twelve markers to track seven segments (feet, legs, thighs, and pelvis). The input information was derived from video recordings (kinematic data from five synchronized cameras and motion analysis using Peak Motus 4.3), force platforms (ground reaction forces), and force transducers mounted on the bed (interaction force between the nurse' leg and the bed or bed reaction force). The author found significant contribution of the contact forces between the bed and the nurses, lower limbs (peak bed reaction forces from 98 N to 222 N) to the L4/L5 moments (from 58 Nm to 160 Nm including the bed reaction forces and from 92 to 227 Nm not including it).

5.4.7. Considerations for data collection in workplace settings

When choosing or developing biomechanical modeling methods for industrial assessments of spinal load, it is important to consider the practicability of the methods within a workplace setting. The time and complexity of instrumenting the worker with any data collection devices, setting up monitoring equipment, and calibrating measures may render the assessment unviable. The expense and availability of equipment and the expertise required to use, maintain, or repair it should also be considered. Equipment should be robust enough to withstand the demands of working environments.

Imaging or joint motion capture systems using video or still cameras are also not without flaws. It can be difficult to get a clear, full body view of a worker performing tasks without interruption workflow. The set up of cameras can be lengthy, but workers may move around the work area or change position relative to cameras so that images are out of plane. Considering the suitability of a data collection method to a specific workplace is just as important as its lab-tested validity, accuracy and reliability. Above all, it is imperative to avoid changing the nature of the work activities when assessing biomechanical loads in workplace settings. If equipment is restricting movement or workers are performing tasks differently, the assessment is unlikely to reflect typical loads.

5.5. Conclusions of Chapter 5

In general, there is a trade-off between the accuracy of a model and the complexity of its calculations or required inputs. The ideal spinal loading model incorporates a balance of precision and convenience. The choice of a biomechanical model depends on the task being assessed, the objectives of the evaluation, and considerations of a model's anatomical realism, and applicability to the situation (e.g. dynamic vs. static; asymmetric vs. symmetric; continuous or not). The loading parameters (e.g. shear, compression) provided by the "ideal" model should provide insights into the WLBD causation process and should be well-related to injury rates. At the same time, the model should require minimal instrumentation especially when large numbers of subjects are to be measured; it should provide enough freedom of movement for the tasks being analyzed, and be inexpensive enough to be used in workplace settings that may impose risk of break downs. Overcoming these challenges will allow for evaluation of lumbar spinal loads and associated risk of WLBD in workplace settings before and after work modifications or training programs using biomechanical models.

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Chapter 6 – Epidemiological and Questionnaire Study of Low Back Disorders in Nurses

This chapter presents the epidemiological and questionnaire study on WLBD in nurses. A version of this chapter has been published in the Journal of Advanced Nursing (Vieira *et al.*, 2006a).

6.1. Introduction

There is a high incidence of WLBD in nurses (Buckle, 1987; Bejia *et al.*, 2005). This is the most prevalent and most costly musculoskeletal disorder (Woolf and Pfleger, 2003). Several epidemiological studies have shown that musculoskeletal disorders (including WLBD) and workload are related (Bernard, 1997). In this paper we present an exploratory study of WLBD and workload from a nursing point of view by administering a questionnaire collecting qualitative information and psychophysical measures of exertion.

Nurses are among those professionals with the highest incidence rates of WLBD (Kumar, 2004). According to Jensen (1987) they have the highest incidence of disabling back disorders in the U.S. The annual incidence rate of WLBD among nurses working in hospitals in France in 1990 was 57% (Niedhammer and Marne, 1994). The incidence of WLBD is also high in Italy (Larese and Fiorito, 1994). In China, the prevalence rate of WLBD in nurses at a teaching hospital was 57% (Smith *et al.*, 2004). WLBD problem is significant worldwide. Hignett (1996) reported an average point prevalence of WLBD of 17 in nurses, an annual incidence rate between 40 and 50, and a lifetime incidence rate between 35 and 80. However, according to the author, these figures are 'a gross under-estimation of the problem' because of the recording systems used and under-reporting (Hignett, 1996, p. 1243). Higher figures were reported by Hofmann *et al.* (2002) in a large cross-sectional questionnaire study: a point prevalence of WLBD of 61 for the nurses; a lifetime incidence rate of 87; and a relative risk between 1.35 and 1.47 for nurses (n = 2207) in relation to administrative clerks (n = 1177). French *et al.* (1997) reported a similar lifetime incidence rate of WLBD of 81 based on a questionnaire study involving 47 Registered Nurses; 92% of nurses who had suffered some form of back pain never reported it.

Bending, twisting, lifting heavy weights and making forceful movements were shown to be related to WLBD (Punnett *et al.*, 1991). Combined lifting, prior injury, and being overweight were found to be risk factors for WLBD among nurses (Fuortes *et al.*, 1994). In a 3-year prospective study including 861 workers, it was seen that spinal kinematics was critical to the development and aggravation of WLBD (Hoogendoorn *et al.*, 2000).

Hignett (1996) did not find any studies using participatory or interview methods to evaluate the problem among nurses, and commented about the limitation and lack of usefulness of the quantitative experimental studies. She reported that qualitative exploratory studies could significantly contribute to the identification of factors leading to the high incidence of WLBD among nurses. Nurses frequently perform tasks involving many of the established risk factors for WLBD, such as patient handling and transfers. Studies of low back pain in nurses are common in the literature, however, the problem is still substantial and little success has been achieved in preventing the problem. For these reasons, it is important to develop assessment tools and implement WLBD control programs in nursing jobs.

The aims of this study were to evaluate the workload, to identify problems that might be leading to the higher incidence rate of WLBD among nurses in the orthopedic and ICU departments of an acuter care teaching hospital in Edmonton, Alberta, Canada, and to gather information about possible improvements that nursing staff would like to see implemented in the workplace. A questionnaire was used to collect the data and its usefulness was investigated.

6.2. Methods

The hospital contacted the researcher because they were forming a "back injury committee" and were interested on prevention programs. The researcher went to the hospital and gave a presentation to the committee members. After the meeting the hospital decided to participate in the research. Initially, a retrospective epidemiological study was performed using information derived from the injury records of the acute care teaching hospital evaluated for the period from 1999 to 2003.

The mean number of workers per year in the hospital in the period was 788 (SD = 174). The recorded cases of WLBD were reviewed, including injury date, job and department where the WLBD occurred, and injury type (first aid, medical aid, or lost time claim). First aid is any work-related injury/illness that did not require healthcare provider treatment and did not necessitate absence from work. Medical aid is any work-related injury/illness that required healthcare provider treatment and did not necessitate absence from work for more than the day of injury. A lost time claim is any work-related injury/illness that necessitated absence from work for more than the day of the injury. Based on the initial evaluation (findings presented in the Results section) it was decided to administer a questionnaire to the Registered and Licensed Practical Nurses of the orthopedics department and of the intensive care unit (ICU) of the hospital. A questionnaire survey was used to collect the data between January and May 2005.

6.2.1. Subjects

All Registered and Licensed Practical Nurses of the orthopedic and ICU departments of the hospital (n = 91) were eligible to complete the questionnaire. The jobs and shifts were the same for all nurses in the orthopedic department, and the ICU had only Registered Nurses. For these reasons, the Registered and Licensed Practical Nurses were evaluated together and are referred to as orthopedic nurses (ONs) or ICU nurses (INs) throughout the paper. Twenty-three ONs (response rate of 96% of the full-time ONs) and 24 INs (response rate of 72% of the full-time INs) completed the questionnaire. Most nurses were female (87% of the ONs, and 96% of the INs). The mean ages for the ONs and INs were 34 (SD = 9) and 36 (SD = 8) years, mean weight was 76 (SD = 6) and 67 (SD = 10) kg, and mean body mass indices were 26 (SD = 5) and 25 (SD = 4) kg/m^2 respectively. The mean height for both groups was 167 (SD = 7) cm.

The response rate (including 98% and 72% of the full-time ONs and INs, respectively) was higher than the minimal (70%) recommended by National Institute for Occupational Safety and Health (NIOSH) for epidemiological studies (Bernard, 1997). Thus, selection bias can be considered negligible.

6.2.2. Nursing job description

The nursing role in the orthopedic department involves medication administration and providing activities of daily living for patients who have undergone orthopedic surgery (e.g. total joint replacements, fractures and back surgery). The activities of daily living help included dressing/undressing, feeding, showering, help in the bathroom and oral care of patients. The job activities also included head-to-toe body assessments, preparing patients for surgery and return from surgery, inserting and removing tubes, catheters and drains. Among the most frequent tasks were: moving and transferring patients; moving furniture and patients from one room to another; turning and repositioning patients in bed and chair; bed making; holding limbs for dressing changes; and office work. The assistance devices most often used for patient transfers were belts, sliding boards and sheets. Mechanical lifts such as a sit-stand, a medi-lift, and a sling lift were also available but were not used as often. The nurses did seven 12-h shifts over 2 weeks (Week 1: 2 days on, 2 days off, 3 days on; week 2: 2 days off, 2 days on, 3 days off). The day and night shift rotation scheme was not clearly defined and varied.

The INs took care of 1 or 2 patients per shift. This involved complete care of critically ill patients who were usually intubated and immobile. The activities included: administering medication, changing intravenous bags and prisma dialysis, monitoring, head-to-toe body assessments, bed bathing, mobilizing, and repositioning patients in bed for procedures and X-rays, adjusting, moving, and lifting equipment. Very frequent activities were lifting patients in the bed (twisting and pulling or pushing the patients – usually unconscious), and turning the patients from side to side (at least every 2 h). The nurses worked four to five 12-h shifts per week, totaling 77.75 h every 2 weeks. The night and day shift rotation was not clearly defined; the nurses worked 2–3 days in a row followed by 2–3 days off.

6.2.3. Questionnaire

The questionnaire was peer-reviewed and published (Vieira *et al.*, 2005). It included questions on the nurses' personal traits, their job characteristics, and on their physical perceptions of their jobs. It also included the following validated psychophysical measures of exertion for the whole job.

Body Part Discomfort Rating (Corlett and Bishop, 1976): This is a body map on which the nurses rated their perceived discomfort by the end of the shift on a 10-point severity scale from ‘no discomfort’ (1), to ‘very uncomfortable’ (10). This technique has been previously validated (Boussenna *et al.*, 1982).

Borg’s Rate of Perceived Exertion (Borg 1962, 1982, 1990): This is a ten-point scale on which the nurses rated their job exertion from ‘nothing at all’ (1), to ‘maximal’ (10). It has been used for more than 40 years, and its validity and reliability have been established by Chen *et al.* (2002), among others. Despite all the changes and advancements over the past 40 years, this tool was included in the questionnaire because it is a validated psychophysical measure of physical stress; it is frequently used by ergonomists, and has been widely accepted in the ergonomics field. For more information on this tool see Dawes *et al.* (2005).

Visual Analogue Scale (Huskisson, 1983): This is a 100-milimeter long horizontal line on which the nurses marked their effort between ‘no effort’ and ‘maximal effort’. The nurses were asked to mark the amount of effort their job requires on five different visual analogue scales (VAS), one for each of the following physical exertion variables – posture, movements, repetitions, force and duration (Vieira *et al.*, 2006c). This tool was initially used to measure pain perception, but it has been used for other types of evaluations and its reliability and validity has been established, including its ability to assess musculoskeletal loads (Huskisson, 1983; Lee *et al.*, 1991).

6.2.4. Ethical considerations and data collection

The study was approved by the university and hospital human research ethics boards. Participation was voluntary; the nurses were informed about the study by a recruitment poster, and those who decided to participate were asked to inform the clinical educator and/or the nurse in charge of their shift. On the days of administration of the questionnaire, the clinical educator was responsible for sending those nurses who had volunteered to participate to a room set aside for this purpose.

The volunteers received further explanations about the objectives and procedures of the study, including a statement of their right not to participate and to withdraw at any time with no consequence to them. After a consent form was signed, the questionnaire was handed out. It was completed during the shift in the presence of the researcher.

6.2.5. Data analysis

The data were analyzed using the software SPSS for Windows. Descriptive statistics were calculated. The annual incidence rate was calculated by dividing the number of new cases of WLBD recorded per year between 1999 and 2003 by the total number of working nurses per year for the same period. The point prevalence rate was calculated by dividing the number of nurses that reported having low back pain at the time of questionnaire completion by the total number of respondents. Finally, the lifetime incidence rate was calculated by dividing the number of nurses who reported having at least one episode of low back injury during their working life by the total number of respondents (Fletcher *et al.*, 1988). The data from the five VASs were added to calculate the total effort required by the jobs, and the total effort was transformed into a percentage of the maximum effort possible (5×100 mm = 500 mm) (Vieira *et al.*, 2006c). The total effort on the VAS and Borg scores were correlated using the Pearson correlation coefficient. Each of the variables on the VAS was standardized as a percentage of the total effort. The discomfort ratings of the different body parts and the five VAS scores were compared using a one-way ANOVA with Fisher's least significant difference post hoc test. The significance level was set to $p < 0.05$.

6.3. Results and Discussion

6.3.1. Retrospective epidemiological study

Between 1999 and 2003, the total number of injuries recorded in the hospital was 677. Of these injuries 547 were first aid (81%), 18 were medical aid (3%), and 112 were lost time claim (16%) injuries. Cases of WLBD represented 23% of all injuries (n = 159), 16% of all first aid (n = 87), 17% of all medical aid (n = 3), and 62% of all lost time claim injuries (n = 69). Seventy-four percent of the WLBD were classified as overexertion injuries (n = 117).

Registered Nurses ($N = 504$) and Licensed Practical Nurses ($N = 96$) had the highest annual incidence rates of WLBD per hundred workers (respectively, 3.1% and 4.3%). The annual incidence rates of WLBD in the nurses were similar to the ones reported previously by Klein *et al.* (1984) for Licensed Practical Nurses (3.3%), and for other heavy jobs involving manual material handling such as lumbermen (3.3%), and construction workers (2.8%). Together, Registered and Licensed Practical Nurses, represented 78% ($n = 123$) of all cases of WLBD recorded in the hospital during the period analyzed and 83% of the lost time claim WLBD cases. Approximately 70% of the WLBD in the Registered and Licensed Practical Nurses happened while transferring or moving patients. The department where the WLBD among Licensed Practical Nurses occurred most often was orthopedics (32%), and the department where the WLBD among Registered Nurses happened most often was the ICU (17%).

6.3.2. Reported lifetime incidence and point prevalence of WLBD

Sixty-five percent of the ONs and 58% of the INs reported to have had at least one WLBD during their working life. While the rate found for the INs was very similar to the 57% found by Bejia *et al.* (2005) in the Fattouma Bourguiba teaching hospital in Monastir, Tunisia, the lifetime incidence rate among ONs was somewhat higher. This trend is also observed when comparing the incidences we found with the 60% 'lifetime prevalence of low back pain' among nurses reported by Smedley *et al.* (1995). Forty-three percent of the ONs and 33% of the INs reported that they were currently experiencing some musculoskeletal pain. Of these, low back pain represented 70% (point prevalence of 30) among the ONs and 75% among the INs (point prevalence of 25). The percentage of low back pain (70% and 75%) among the nurses that reported some type of musculoskeletal pain currently, was higher than the 63% previously reported by French *et al.* (1997) among Registered Nurses ($n = 47$) in a similar study. The orthopedics departments of hospitals were previously found to have a higher incidence of WLBD in relation to other departments (Yassi *et al.*, 1995). In addition, the WLBD point prevalence found for the ONs (30) was higher than those found for different populations. Reigo *et al.* (1999) found a back pain point prevalence of 23 [95% confidence interval (CI) 21–25] for the Swedish population aged between 20 and 59 years.

Picavet and Schouten (2003) found a back pain point prevalence of 27 (95% CI 26–28) for the Netherlands population. Likewise, Walker *et al.* (2004) found a back pain point prevalence of 26 (95% CI 24–28) for the Australian population. The sample size (n = 23 ONs and 24 INs) is too small too allow statistical comparisons between the groups, and any generalizations to the general population of nurses would be inappropriate. Despite this limitation, the data show some interesting trends within the hospital; the data also concur with other studies using bigger samples.

6.3.3. Exercising, smoking and WLBD

The ONs and INs who reported that they exercised regularly (respectively, 57% and 50%) said that the mean time they exercised was 198 (SD = 119) and 195 (SD = 93) minutes/week, respectively. The ONs and INs who reported that they smoked (respectively, 35% and 21%) said that they smoked 8 (SD = 5) and 10 (SD = 3) cigarettes per day, respectively. Eighty percent of the ONs and 70% of INs who exercised regularly did not report low back disorders; neither did 80% of the ONs and 90% of the INs who did not smoke, and 67% of the ONs and INs who exercised and did not smoke. On the other hand, 62% of the ONs and 64% of the INs who did not exercise reported low back disorders, and so did 75% of the ONs and 80% of the INs who smoked, and all ONs and INs who did no exercise and smoked.

These findings point to a relationship between smoking and lack of exercise with low back disorders (Vieira *et al.*, 2006b). Despite the small sample size of this study, the observed trend is in agreement with other studies. Tsai *et al.* (1992) found higher OR of low back disorders for smokers (1.54, $p < 0.01$) and overweight workers (1.42, $p < 0.01$). Bejia *et al.* (2005) found a WLBD OR of 1.69 (95% CI = 1.03–3.07) for tobacco users and also found that exercising has a protective effect ($p = 0.019$). According to Frymoyer *et al.* (1983), the cigarette nicotine causes vasoconstriction, which reduces the blood flow to the muscles and the intervertebral discs, and predisposes smokers to low back disorders. In a previous study, the authors suggested that increased coughing among smokers is also related to increased risk of low back disorders in this group (Frymoyer *et al.*, 1980).

Additionally, Fogelholm and Alho (2001) proposed the hypothesis that cigarette smokers have an increased risk of low back disorders because of intervertebral disc degeneration and spinal instability caused by increased proteolytic activity. The authors cite studies where the OR of low back disorders among smokers compared with non-smokers varies between 1.3 and 2.5. The percentage of the ONs who reported that they smoked (35%) is higher than that of the province of Alberta or the Canadian national averages, which are, respectively, 23% and 22% of the population (Bowerman, 2004). In addition to smoking and a sedentary life style, advanced age, female gender, increased body mass index, and disturbed psychological profile are among other individual factors shown to be related to an increased risk of low back disorders (Bejia *et al.*, 2005).

6.3.4. Manual handling

The weight handled and the numbers of transfers by the INs were smaller than those reported by the ONs ($p < 0.006$). The number of lifts in general (patients, furniture, or equipment) reported by the INs was also smaller ($p = 0.029$); the differences between the number of lowers, pushes, and pulls during the work shift were not statistically different (Table 6-1).

Table 6-1. Mean (SD) weight handled, number of transfers and number of repetition of tasks per shift by orthopedic (ONs) and intensive care nurses (INs).

	Weight handled in kg	Number of transfers performed			Number of repetitions of the tasks			
		bed to chair/ commode	stretcher to bed	chair to chair/ commode	lifting	lowering	pulling	pushing
ONs	47 (30)	11 (6)	8 (5)	4 (2)	19 (13)	11 (9)	9 (6)	9 (7)
INs	26 (10)	1 (1)	1 (1)	0 (1)	13 (8)	8 (5)	14 (11)	12 (8)

The difference between the weights manually handled by the ONs and INs may be explained by the smaller number of transfers performed by the INs. However, the INs reported that they turned their usually incontinent patients from back to side and vice versa every two-hours during the shift and also helped the other INs to turn their patients. The mean number of patient turns in bed per shift reported by the INs was 8 (SD = 3). The fact that the INs perform less patient transfers (lifting and lowering), but more patient turns (pulling and pushing) is reflected by the number of repetitions of the tasks.

The variation on the average weight and the number of transfers and repetitions of the tasks reported by the nurses may be explained by the variability in the job itself. Despite the differences, the weight manually handled, as reported by both ONs and INs, was higher than the limit proposed in the NIOSH 1991 lifting equation as the maximum recommended weight (23 kg) for occasional lifting in the sagittal plane, with good couplings, and vertical displacement of less than 25 cm (Waters *et al.*, 1993). Manual transfer and repositioning patients in bed were previously reported as being associated with an increased risk of WLBD (Smedley *et al.*, 1995). Manual patient transfer was reported by nurses as the most stressful method (Garg and Owen, 1994).

6.3.5. Very frequent postures and activities in the jobs

Figure 6-1 presents the percentage of the nurses who marked different postures as very frequent in their job. Standing with the trunk flexed was the posture most frequently identified as very frequent in the job of both ONs and INs.

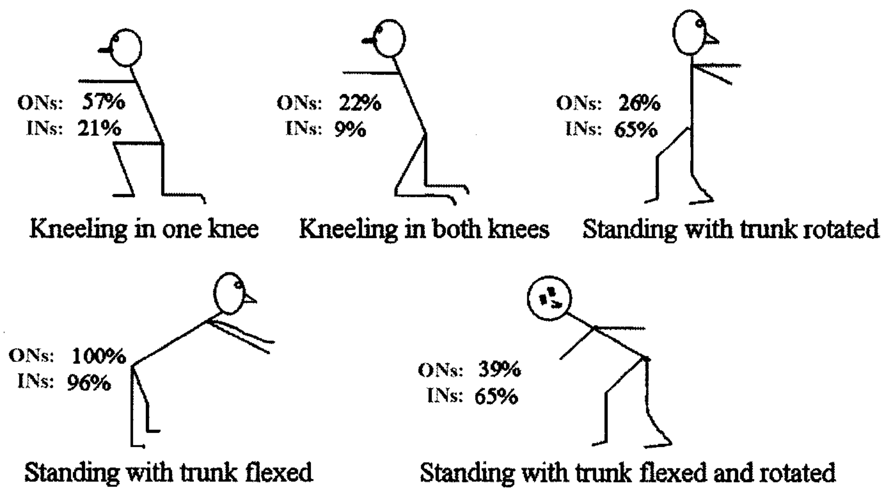

Figure 6-1. Percentages of orthopedic nurses (ONs) and intensive care nurses (INs) who classified different postures as 'very frequent' in their jobs.

In addition to the postures in Figure 6-1, 32% of the ONs and 22% of the INs reported other postures. Squatting (63% of the ONs and 40% of the INs), and twisting and reaching (37% of the ONs and 60% of the INs) were also very frequent postures.

The mean time the nurses reported that they maintained a static posture during an activity before a break or change in position was 2.4 (SD = 1.7) minutes for the ONs and 4.1 (SD = 2.2) minutes for the INs. Squat lifting and pulling were the activities most often identified as very frequent in the job of an ON, while for the INs the most frequent activities were pushing and pulling. These findings agree with the reported numbers of the tasks (Table 12) and show the internal consistency of the questionnaire. Other activities mentioned were holding a limb up, and combined lifting and pulling by the ONs, and bringing the patient up in bed by the INs.

6.3.6. Reported discomfort by the end of the shift

Figure 6-2 presents the discomfort experienced by the end of the shift rated by the nurses on the body part discomfort index (ten-point scale).

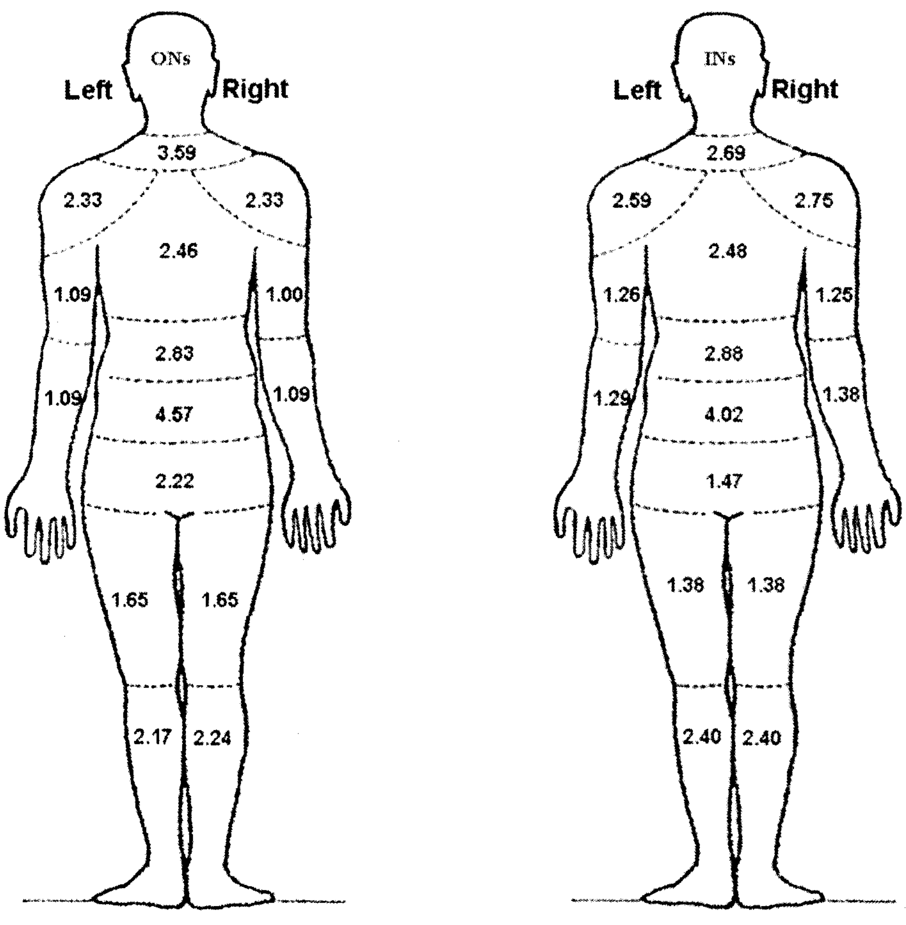

Figure 6-2. Discomfort felt by the end of the shift in different body parts by orthopedic (ONs) and intensive care nurses (INs) – 10-point scale.

In ONs the low back discomfort rate was higher than for most other body parts ($p < 0.01$) except for the neck ($p = 0.06$). In INs, the low back discomfort rate was higher than for all the body parts ($p < 0.03$). There were no significant differences between the discomfort rates indicated for each of the body parts by ONs and INs.

6.3.7. Rates of perceived exertion

The perceived exertion on the 10-point Borg scale was 6.7 (SD = 1.8) (very strong) by the ONs and as 5.8 (SD = 1.9) (strong) by the INs. On the VAS, the total effort required by the jobs was 67% (SD = 14) of the maximum for the ONs, and 68% (SD = 15) for the INs. There were no differences between the scores given by the ONs and INs to each variable on the VAS (posture, movements, repetitions, force and duration), and no variable scored significantly higher than the others. Figure 6-3 presents the results of the visual analogue scales. The VAS scores (0 to 100 each) of the five VAS filled up were summed (maximum sum = 500), this total score was considered to be 100% and the respective percentages represented by each variable were calculated [variable % = (100 x variable score/total score)].

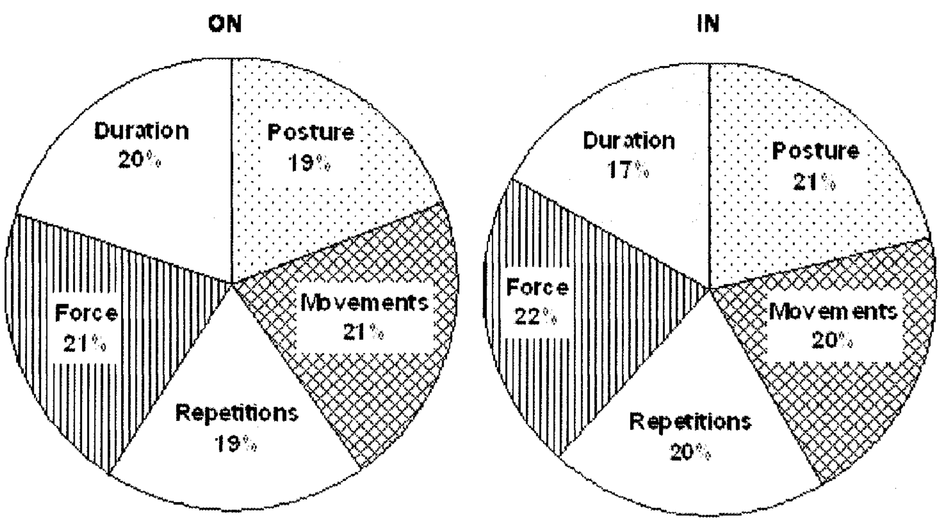

Figure 6-3. Contributions of posture, movements, repetitions, force, and duration to the total effort required by the roles of orthopedic (ONs) and intensive care nurses (INs) (physical effort rated using Visual Analogue Scales).

Because of the fact that five physical exertion variables were evaluated as representing the total job effort, a 20% representation for each would be expected if the physical workload was equally distributed (Vieira *et al.*, 2006c).

Even though no significant differences were found, in general, force and movements contributed more to the total ONs job effort, while force and posture contributed more to the total INs job effort. This trend is in agreement with the observed job demands and described job tasks. Thus, this method of analyzing the physical effort in relation to its different components seems to be interesting and useful. Additional research on its use and applications should be carried out.

6.3.8. Problems related to WLBD and suggested improvements

Table 6-2 presents the issues reported as related to the high incidence of WLBD among the nurses and their suggestions for improvement.

Table 6-2. Problems of and suggestions by orthopedic (ONs) and intensive care nurses (INs) to reduce the number of work-related low back disorders in their jobs.

Problems	%*		Relevant suggestions	%*	
The rooms are too small	29 6	ONs INs	Bigger rooms and to adjust environment (rooms) set up ergonomically	13 6	ONs INs
Staff shortage and lack of encouragement to ask for help	15 39	ONs INs	Increase staff and/or add assistants to help with transfers and encouragement to ask other staff members (nurses, physiotherapists, occupational therapists) for help and to use assistance devices even if it takes more time	28 25	ONs INs
Patients too heavy for the staff and unexpected movements	15 14	ONs INs	Pre-employment functional capacity evaluations and regular exercise program	5	ONs
Working with inexperienced staff; lack of training on lifting and patient transfer, and the postures required	13	ONs	Training in lifting and patient transfer, especially for new staff, and to be reminded to use proper posture	10 10	ONs INs
Manually handling heavy bags. The nurses have to squat to hang 5 kg bags of fluid on the holders of the dialysis machine, and some of the intravenous infusion poles are not adjustable	13	INs	New adjustable intravenous poles and higher fluid bag holders on the dialysis machines	9	INs
Shortage of lifting devices for moving and patient transfers; equipment is less than ideal; it is too time consuming, or the equipment is not well maintained	10 6	ONs INs	Additional and more adequate devices for patient transfer	18 31	ONs INs
Many transfers due to surgeries, X-rays, and room transfers, and having to rush to get next patient discharged or admitted from surgery	8	ONs	Rotate the staff due to the high acuity of the patients and put very heavy patients in different rooms, and to spread the number of X-rays, surgeries, tests throughout the week to avoid peak days	8 3	ONs INs
The X-rays are done at 06.00 in the ICU. At this time the nurses are tired because it is close to the end of the shift.	13	INs	Beds with X-ray capabilities and schedule the X-rays for the beginning of the shift	13	INs
Shifts are too long (12 h)	6	ONs	Reduce shifts to no longer than 8 h	8	ONs
Repetitive holding of patients' limbs and reaching over the bed to reposition the patients	6	INs	Always encourage patient to help as much as possible	8	ONs
Overestimation of patient capabilities and lack of patient education pre-operation	2	ONs	Give guidelines for lifting/turning patients	3	INs
Inadequate system to lower bed rails	2 3	ONs INs	Patients should come back from operation room already in their beds	2	ONs

* – percentage of the responses by the ONs and INs

Lifting and transferring patients was previously reported as the main cause of WLBD among nurses (Yassi *et al.*, 1995). Stobbe *et al.* (1988) compared the frequency of WLBD in two groups. One group was classified as having high frequency of patient lifting (more than five patient lifts/shift, $n = 317$) and the other group was classified as having low frequency of patient lifting (less than two patient lifts/shift, $n = 98$). Their study showed that lifting patients was directly associated with the probability of nurses experiencing WLBD; the low frequency of lifting group 'survived' longer without a WLBD than the high frequency group (10% difference after 1215 days, $p < 0.01$). The Licensed Practical Nurses in the high lifting frequency group were 7.54 times more likely to have a WLBD than those in the low frequency of group (Stobbe *et al.*, 1988).

In addition to the well recognized and important relationship between patient manual handling and WLBD, the results presented in Table 13 also point out other problems related to WLBD in nursing jobs that deserve attention. Among these problems it is relevant to highlight environmental and organizational issues such as the room sizes and setup, and the X-ray and shift schedules. Training in biomechanics and lifting techniques helps, but it alone is not enough to reduce the number of cases of WLBD in the workplace (Hignett, 1996). The introduction of new assistance devices results in higher compliance to prevention programs than training alone (Daynard *et al.*, 2001). Marras *et al.* (1999) reported that manual patient handling is an extremely risky task for WLBD. Elford *et al.* (2000) studied patient transfer from chair to chair with and without lifting slings. The authors found higher velocities, accelerations, and body part stress reported on VAS for the manual transfers than when using lifting slings. However, not all studies show that the use of lifting devices for patient transfers is always good. The main problem is that transfers take longer when using assistance devices (Garg and Owen, 1994). Daynard *et al.* (2001) showed that, despite reducing the peak load, the use of some devices may increase the total cumulative load because of the longer time required to perform the transfers. Cumulative compression and shear forces were shown to be higher in institutional aids with low back pain than in those without pain (Kumar, 1990). Thus, each situation has to be studied specifically in order to find the safest way of moving and transferring patients.

Ergonomic assessment and interventions have been shown to improve the control of musculoskeletal disorders among nurses (e.g. Garg and Owen, 1994). Hignett (2001) reports a successful ergonomic experience in a Hospital in UK. The five-year program resulted in 33% reduction in manual handling incidents, and 36% reduction in days lost through musculoskeletal disorders. A generally accepted approach is to identify tasks with high risk, determine the critical factors in the tasks, and modify those factors to reduce the risk of WLBD. Different modifications are being discussed based on the results. In general, additional equipment for patient transfer, training programs, bigger rooms, adequate room set up, and additional staff are recommended interventions. Considering that the questionnaire response rate was higher than 70%, the results and conclusions can be generalized for all the ONs and INs of this hospital (Bernard, 1997). However, the sample size was small and the results and conclusions should not be generalized for nurses in every hospital. Specific studies are required in different hospitals because the problems potentially causing WLBD may differ. The questionnaire used was an adequate tool for identifying the relevant issues that should be addressed.

6.4. Conclusions of Chapter 6

The questionnaire used is practical and useful to initiate an evaluation of nursing jobs. It facilitates the systematic collection of staff input. It helps to identify problems and possible solutions or improvements to nursing jobs that can be used to design participatory ergonomic interventions aimed at reducing WLBD. The jobs in the orthopedic and ICU departments have heavy workloads. Many WLBD risk factors such as bending, twisting, lifting heavy weights and making forceful movements are present in these jobs. However, there are differences between the jobs. Thus, nursing jobs should be evaluated specifically according to the department because the demands and risks differ. Specific modifications are being studied. The general recommendations to improve the control of WLBD in the jobs are:

1. Increasing the number of patient transfer devices;
2. Delivering training programs;
3. Increasing the size of the rooms and/or adjusting the room set-up ergonomically;
4. Hiring additional nurses and/or nursing aids to help with patient turns and transfers.

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Chapter 7 – Epidemiological and Questionnaire Study of Low Back Disorders in Steel Workers

This chapter presents the epidemiological and questionnaire study on WLBD in steel workers. A version of this chapter has been submitted for publication and is under review in the International Journal of Industrial Ergonomics (Vieira and Kumar, 2006b).

7.1. Introduction

Work-related low back disorders (WLBD) are common and represent the most costly musculoskeletal disorder (AHRE, 2003; BLS, 2001; Nordin *et al.*, 1997). WLBD are frequent in the steel industry (Dueker *et al.*, 1994). For example, Udo and Yoshinaga (2001) reported that about 70% of the workers of the maintenance division of a steel mill had back pain during life. Masset and Malchaire (1994) studied WLBD in two steel companies using a checklist. The prevalence of low back pain was 66% for lifetime, 53% during previous year, and 25% during the previous week. Similarly, Hildebrandt *et al.* (1996) found a prevalence of low back symptoms in steel workers of 53% after one year of work.

WLBD control is an important issue in occupational health. Several epidemiological studies have demonstrated evidence that musculoskeletal disorders and workload are related (Bernard, 1997; Hales and Bernard, 1996). Bending, twisting, lifting heavy weights, and making forceful movements were shown to be related to WLBD (Frymoyer *et al.*, 1980; Punnett *et al.*, 1991). Welders and computer numeric control (CNC) workers frequently perform tasks involving the established risk factors for WLBD. For these reasons, it is important to implement WLBD control programs in welding and CNC.

The cause of work-related musculoskeletal disorders is multifactorial (Kumar, 2001). We recently published the results of a questionnaire study on the relationship between personal risk factors (smoking, non-exercising, and being overweight) and WLBD in steel workers and nurses (Vieira *et al.*, 2006b). In addition to the personal risk factors, the occupational risk factors are also an important aspect to consider on the control of WLBD.

Perceived workload is an invaluable means to assess the occupational risk factors for WLBD, and the workers expertise is helpful in identifying suitable intervention measures (Hignett, 1996 and 2001; Hildebrandt *et al.*, 1996; Udo and Yoshinaga, 2001; Vieira *et al.*, 2006a). For these reasons, the objectives of this paper were (I) to present the results on the perceived occupational risk factors (perceived workload and reported physical demands) by welders and by computer numeric control (CNC) workers in two steel companies, (II) to report the problems identified by the workers as potential causes of WLBD in their jobs, and (III) to introduce the potential improvements suggested by the workers to reduce the incidence of WLBD in their jobs. For the nurses, the information on these aspects was presented elsewhere (Vieira *et al.*, 2006a).

7.2. Methods

The steel companies participating in the study were informed about the project by the Manufacturers' Health and Safety Association of Alberta (MHSA). The MHSA was initially contacted by the researcher who told the director of the association about the study and asked him to inform steel companies with high incidence rates of WLBD about the study and ask them to contact the researcher in case they were interested. The two steel companies in the study contacted the researcher saying that they wanted to know more about it. The researcher went to the companies and gave a presentation to the manufacturing and health and safety management staff. After the meeting the companies decided to participate in the study.

7.2.1. Epidemiological study

A retrospective study (1999-2003) of WLBD claims (low back disorders registered by a workers compensation agency) was conducted in steel companies A and B. The epidemiological evaluation was performed using previously recorded information from the Workers Compensation Board and from internal records of injuries for these worksites. Only new cases were considered, the re-injuries were not included in the calculations. In addition, the worksites were asked to provide information about the number of workers per job per year from 1999 to 2003. The WLBD claims were reviewed, including injury date and job of the injured workers.

7.2.2. Questionnaire survey

After the retrospective study, the welders of company A and the CNC of company B completed a questionnaire (Vieira *et al.*, 2005 and 2006a). The questionnaire included questions on the workers' personal traits, life style, history of WLBD, perceived occupational factors, and the following scales.

10-point Body Part Discomfort Rating: A body diagram on which the workers identified and rated areas of perceived discomfort by the end of the shift on a 10-point severity scale from 'no discomfort' (1) to 'very uncomfortable' (10) (Corlett and Bishop, 1976).

10-point Borg's Rate of Perceived Exertion: A ratio-categorical scale on which the workers rated their job exertion from 'nothing at all' (1) to 'maximal' (10) (Borg, 1962, 1982, 1990).

100-milimeter Visual Analogue Scales (VAS): Horizontal lines on which the workers marked their job required efforts for posture, movements, repetitions, force, and duration between 'no effort' and 'maximal effort' (Huskisson , 1983; Vieira *et al.*, 2006c).

7.2.3. Sample

One-hundred and eight male workers (64 welders and 44 CNC workers) from the steel companies A and B completed the questionnaire. The sample included, respectively, 78% (64/84) and 94% (44/45) of the full time welders and CNC workers in these worksites. The participation rates were more than the minimal (70%) recommended by NIOSH (Bernard, 1997). The sample characteristics are presented in Table 7-1.

Table 7-1. Mean (SD) of age, height, weight, and body mass index of the welders and computer numeric control (CNC) workers.

Job	Age (years)	Height (cm)	Body mass (kg)	Body mass index (kg/m^2)
Welders (n = 64)	37 (13)	177 (10)	81 (18)	26 (5)
CNC (n = 44)	27 (5)	178 (1)	83 (13)	26 (4)
Total (n = 108)	33 (11)	177 (9)	82 (16)	26 (5)

There were no differences between the welders and CNC workers in height, weight, and body mass index (BMI). However, the welders were on average 10 years older than the CNC workers ($p < 0.001$, 95% CI 6 to 14).

The BMI is a measure of body fat based on height and weight. Depending on the value a person is considered underweight (BMI < 18.5), normal weight (BMI from 18.5 to 24.9), overweight (BMI from 25 to 29.9), or obese (BMI ≥30) (National Heart, Lung, and Blood Institute, 2005). Thus, both welders and CNC workers were overweight.

7.2.4. Jobs

Steel company A manufactures structural steel for the construction and oil industries. Steel pieces welded at steel company A included beams, plates, frames, and structural steel components. The welders received the pieces, inspected, lifted and carried smaller pieces from stacks to workbenches approximately 3 m apart, welded pre-fitted joints on handrails and structural pieces using wire weld, grinded and air-chipped welds, replaced wire reel and performed housekeeping.

Larger pieces were moved using cranes and frequently caused the workers to assume awkward postures to get to the joints to weld, sometimes in small and confined spaces. After the job was completed, the pieces were moved to sand blasters. The work shift was 10 h, 4 to 5 days/week; there was monthly rotation between day and night shifts. The day shift was from 5:30 am to 4 pm with 30 min lunch from 10:30 am to 11 am, and the night shift was from 4:30 pm to 3 am with 30 min diner from 9:30 pm to 10 pm. Figure 7-1 shows a welding job in kneeling posture.

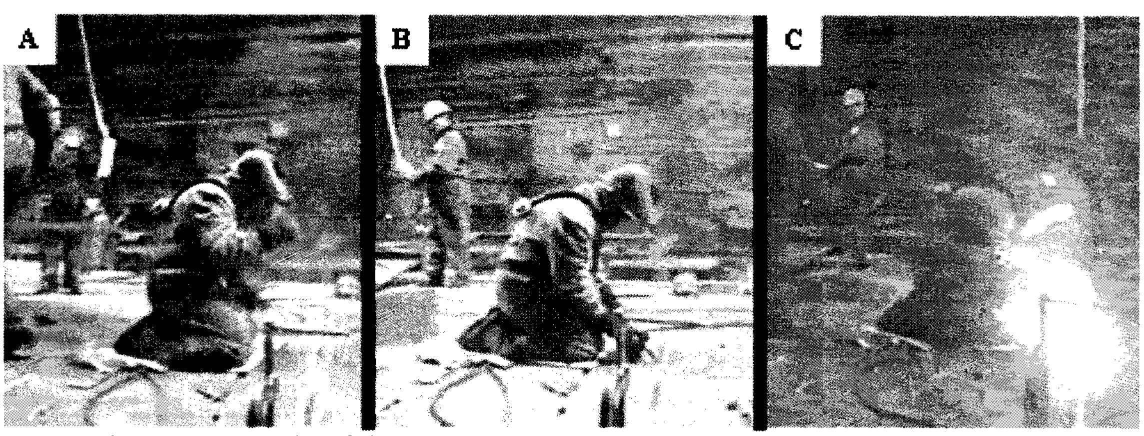

Figure 7-1. Cycle of the welding job, sequence left (A) to right (C).

Steel company B processed blades for tractors, snowplows and other machines. The CNC job at company B required to program, set clamps, load, tighten up and untighten blades to machines, unload machines manually or using cranes, inspect and measure blades, and grind edges and holes. When using cranes, a magnet was attached to the blade and it was pushed or pulled towards the table, bench, or blocks on the floor. The blades were pushed or pulled on rollers while on machine tables. The machines drilled, milled, or punched holes. The blades were flipped over for cleaning the shavings out then stacked and moved to the next station. At the end of the shift, the workers cleaned-up shoveling out the shavings. The work shift was 12 h/day, 42 h/week; there was monthly rotation between day and night shifts. Figure 7-2 shows a CNC worker loading and unloading a drilling machine.

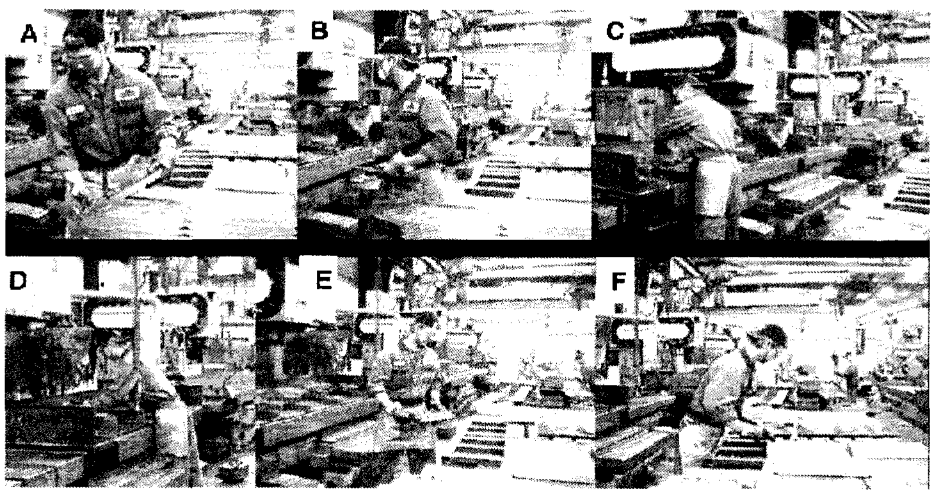

Figure 7-2. Cycle of the computer numeric control job, loading and unloading a drilling machine, sequence from top-left (A) to bottom-right (F).

7.2.5. Data collection and ethics

This study was approved by the University Human Research Ethics Board. Participation was voluntary. The workers who decided to participate were sent them to fill out the questionnaire. The questionnaires were completed during the shifts at the worksites in the presence of the researcher. The volunteers received explanations about the study objectives and were informed of their right not to participate and to withdraw at any time with no consequence to them. All questions were answered, and the questionnaires were handed out after a signed informed consent was obtained.

7.2.6. Data analysis

There is some confusion on the use of the epidemiological terms of incidence and prevalence. In this study we use the following definitions: (I) annual incidence is the number of new cases recorded in a year divided by the number of workers in that year multiplied by 100; (II) working-life incidence is the number of workers that reported to have had at least one episode during their working-life divided by the number of workers questioned multiplied by 100; (III) point prevalence is the number of workers reporting to have the condition at that moment divided by the number of workers questioned multiplied by 100 (Portney and Watkins, 2000). The annual incidences of WLBD claims (average percentage of workers that had a WLBD claim recorded by the workers compensation board between 1999 and 2003) were calculated dividing the numbers of claims per year (excluding re-injuries) by the number of workers per year. Similarly, the working-life incidences of WLBD (explained for the workers as 'low back pain interfering with daily activities and work during the working-life') and point prevalences of low back pain (percentage of the workers that reported to suffer pain on the lumbar spine region at the moment of questionnaire completion) were calculated dividing the frequency by the number of workers that completed the questionnaire.

The five VAS results were added to calculate the total effort required by the jobs. Each VAS was presented as a percentage of the total effort and the total effort was presented as a percentage of the maximum possible (500 mm). The data analysis incorporated descriptive statistics (measures of central tendency and variability) and comparison of means using the SPSS statistical package for windows (SPSS Inc., Chicago, IL). Numerators and denominators were given in parenthesis for percentages. The homogeneity of variances was analyzed using Levene statistic. The sample characteristics (age, height, weight, and BMI) were compared using independent-samples t-test. The body parts discomfort ratings, and the five VAS, were compared using one-way ANOVA with Fisher's least significant difference post-hoc test. Confidence intervals (95% CI) were presented for significant ($p < 0.05$) differences.

7.3. Results

7.3.1. Frequency of WLBD in the welders and CNC workers

In total (including re-injuries), there were 37 WLBD claims, from 1999 to 2003, among the 144 workers in the blue collar jobs in company A (welding, fitting, and CNC), and 39 WLBD claims in the 95 workers in the blue collar jobs in company B (CNC, cutting, and heat treatment). The workers with the highest number of WLBD claims in company A were welders [38% (14/37)], fitters [16% (6/37)], and fitter apprentices [11% (4/37)], while in company B they were CNC workers [44% (17/39)], cutting workers [36% (14/39)], and heat treatment workers [21% (8/39)]. The direct cost of WLBD claims was C$ 24 471 to company A and C$ 69 997 to company B.

The annual incidences of WLBD claims was 3.4% (1.85/55) for the welders and 5.4% (2.17/40) for the CNC workers; the working-life incidences of WLBD were 55% (35/64) for the welders and 36% (16/44) for the CNC workers, and the point prevalences of low back pain were 27% (17/64) for the welders and 16% (7/44) for the CNC workers. Musculoskeletal pain in some body part at the time of questionnaire completion was reported by 44% (28/64) of the welders and 27% (12/44) of the CNC workers. From those, 61% (17/28) and 58% (7/12) respectively reported low back pain. Thus, the point prevalence of musculoskeletal pain in the total sample was 37% (40/108).

7.3.2. Body Part Discomfort Rating by the end of the shift

Figure 7-3 presents the discomfort by the end of the shift in different body parts. The welders presented significantly higher scores than the CNC workers for discomfort at the neck, upper arms, low back, buttocks, and thighs. There were differences between the discomfort scores given to the different body parts by both welders and CNC workers (Table 7-2). The scores given to the different body parts were compared to each other (comparison between the body parts) within each group (welders and CNC workers).

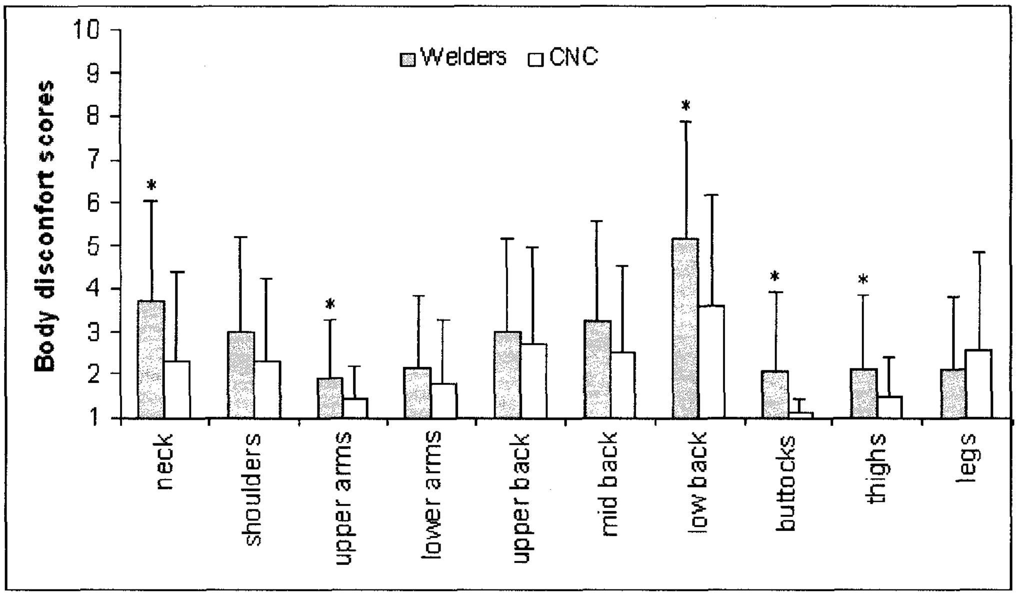

Figure 7-3. Discomfort scores (mean and SD) in different body parts by the end of the shifts: comparison between welders (n = 64) and computer numeric control workers (CNC, n = 44) (1 = no discomfort; 10 = maximum discomfort; * = $p < 0.04$).

Table 7-2. Analysis of variance (one-way ANOVA) among the discomfort scores given for different body parts by welders and computer numeric control workers (CNC).

Workers	ANOVA	Sum of Squares	df	Mean Square	*F*	*p*
Welders N = 64	Between Groups	599.743	9	66.638	15.726	< 0.001
	Within Groups	2669.675	630	4.238		
	Total	3269.418	639			
CNC N = 44	Between Groups	217.346	9	24.150	7.393	< 0.001
	Within Groups	1404.676	430	3.267		
	Total	1622.022	439			

The low back received higher discomfort scores (verbal correspondent of 'moderate discomfort') than all other body parts for both welders and CNC workers (Table 7-3).

Table 7-3. Fisher's least significant difference post-hoc test between the low back discomfort scores and the scores given to other body parts by welders and computer numeric control workers (CNC).

Workers	low back vs.	Mean Difference	Std. Error	*p*	95% Confidence Interval Lower Bound	95% Confidence Interval Upper Bound
Welders n = 64	buttocks	3.094	0.364	< 0.001	2.379	3.808
	legs	3.047	0.364	< 0.001	2.332	3.761
	lower arms	3.008	0.364	< 0.001	2.293	3.722
	mid back	1.891	0.364	< 0.001	1.176	2.605
	neck	1.438	0.364	< 0.001	0.723	2.152
	shoulders	2.129	0.364	< 0.001	1.414	2.844
	thighs	3.047	0.364	< 0.001	2.332	3.761
	upper arms	3.234	0.364	< 0.001	2.520	3.949
	upper back	2.156	0.364	< 0.001	1.442	2.871
CNC n = 44	buttocks	2.489	0.385	< 0.001	1.731	3.246
	legs	1.023	0.385	0.008	0.265	1.780
	lower arms	1.784	0.385	< 0.001	1.027	2.541
	mid back	1.057	0.385	0.006	0.299	1.814
	neck	1.273	0.385	0.001	0.515	2.030
	shoulders	1.273	0.385	0.001	0.515	2.030
	thighs	2.114	0.385	< 0.001	1.356	2.871
	upper arms	2.170	0.385	< 0.001	1.413	2.928
	upper back	0.898	0.385	0.020	0.140	1.655

7.3.3. Manual materials handling and rates of perceived physical exertion

Table 7-4 presents the number of times per hour that the welders weld, grind, and chisel, and that the CNC workers lift, push and pull during the shift.

Table 7-4. Frequency per hour [mean and standard deviation (SD)] of common tasks in the welding and computer numeric control (CNC) jobs.

	Welders (n = 64)			CNC (n = 44)		
	Welds	Grinds	Chisels	Lifts	Pushes	Pulls
Mean	24	10	20	18	18	18
SD	16	11	16	15	15	15

Table 7-5 presents the reported weights handled, the perceived exertion scores (Borg scale) and the total effort required (VAS) by the jobs.

Table 7-5. Mean (SD) of the reported weights manually handled, the perceived exertion rated in the Borg scale, and the total effort required in visual analogue scales rated by welders and computer numeric control workers (CNC).

Workers	Weight (kg)	Perceived Exertion (10-point scale)	Effort required (% of maximum)
Welders (n = 64)	21 (6)	5 (1)	62 (14)
CNC (n = 44)	35 (11)	5 (1)	62 (11)

The perceived exertion rated on the Borg's 10-point scale was strong for both groups. The reported weights handled by the welders was lower than by the CNC workers, but the perceived exertion and the total effort required were not different between the groups (Table 7-6).

Table 7-6. Independent samples t-test for equality of means of the weight handled, the perceived exertion, and the effort required by the welders (n = 64) and computer numeric control workers (n = 44).

Variable	*t*	df	*p* (2-tailed)	Mean Difference	Std. Error Difference	95% CI of the Difference	
						Lower	Upper
Weight Handled	-7.297	98	<0.001	-14	2	-10	-18
Perceived Exertion	1.356	100	0.178	0.4	0.3	-0.2	0.9
Effort Required	0.123	101	0.902	0.3	2.5	-4.6	5.2

The total effort required by the jobs on the VAS for welding and CNC were not significantly different. The maximum possible was 500 mm representing the sum of the five 100 mm VAS for postures, movements, repetitions, force, and duration. The total score was considered to be 100% and the respective percentages represented by each variable were calculated [variable % = (100 x variable score/total score)]. Figure 7-4 presents the percentages of contribution of the different variables to the total effort required by the jobs.

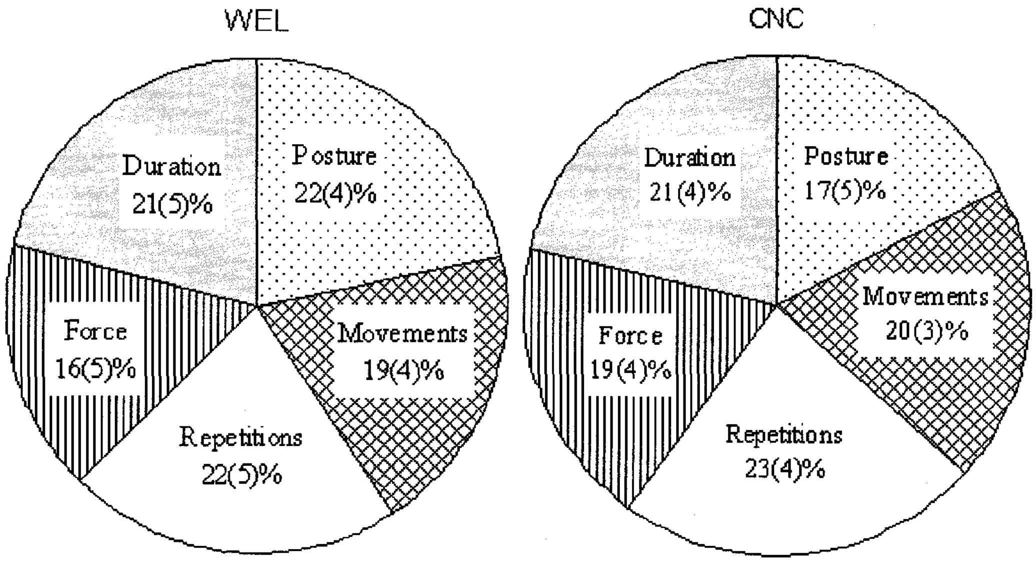

Figure 7-4. Mean (SD) percentages of contribution on visual analogue scales for the postures, movements, repetitions, forces and durations of the job to the total job effort by welders (WEL, n = 64) and computer numeric control workers (CNC, n = 44).

For the welders, the postures, repetitions, and duration contributed more to the total effort than the movements and forces ($p < 0.035$, 95% CI 0.2 to 7.0). For the CNC workers, repetitions and duration contributed more to the total effort than postures, movements, and force ($p < 0.044$, 95% CI 0.1 to 8.0). The contribution of postures to the total effort by welders was higher than for CNC workers ($p = 0.001$, 95% CI 2.8 to 6.4). On the other hand, the contribution of forces was lower for welders than for CNC workers ($p = 0.034$, 95% CI -3.9 to -0.2).

7.3.4. Problems related to and suggested improvements to reduce WLBD

Table 7-7 presents the issues reported as related to WLBD in the jobs and the workers' suggestions.

Table 7-7. Problems of and suggestions by welders and computer numeric control workers (CNC) to reduce WLBD in their jobs.

Workers	Problem	Suggestion
Welders	Awkward postures	Improve the workstation set up and space. Reduce the size and weight of the welding wire rolls. Rotate jobs to avoid prolonged exposure to awkward postures. Minimize the time in awkward postures. Change postures when possible. Plan task before execution.
	Low welding tables	Height adjustable welding tables.
	Time and production pressures	No suggestions from workers.
Both	Behavior and education	Biomechanics education and training / lifting techniques. Exercising/stretching programs before starting and during the job. Use crane to move and re-position heavy material.
	Moving the cranes with heavy loads	Maintain the cranes in good working condition. Reduction of crane momentum.
	Lack of clear standards and their enforcement	Clear lifting limits.
CNC	Standing on bare concrete	Installation of fatigue resistant mats.
	Cleaning machines at the end of the shift (lack of space at the back / awkward postures)	Reposition the machines to provide easy access to all parts including the back.
	Asymmetric lifting	Reduction of asymmetrical lifting.
	Tables on machines are too deep (bending and reaching for loading and unloading)	No suggestions from workers.
	New employees start as punch operators ('the punch operator is continuously pushing, pulling, and lifting blades')	No suggestions from workers.

7.4. Discussion of Chapter 7

7.4.1. Incidences and prevalences

The annual incidence of WLBD claims for the welders (3.4%) was similar to, while the annual incidence for the CNC workers (5.4%) were somewhat higher than, those reported for licensed practical nurses (3.3%), lumbermen (3.3%) and construction workers (2.8%) (Klein *et al.*, 1984). A 2% annual incidence of WLBD claims was estimated for the employed American population (Kelsey and White, 1980). Similarly, a 2% annual incidence of WLBD resulting in work absence was found for male workers between 25 and 44 years old from Quebec (Canada) (Abenhaim and Suissa, 1987). Thus, the incidence of WLBD in the welders, and especially in the CNC workers, can be considered high.

The working-life incidence rate of WLBD for both welders (55%) and CNC workers (36%) groups were lower than the 'lifetime' incidence of 66% previously reported for steel workers (Masset and Malchaire, 1994). In this study, we asked if the workers had low back pain that interfered with daily activities and work during their working-life, whereas the previous study reported low back symptoms during lifetime. Reported lifetime incidence of 'low back pain' in the US population varied from 48.8% to 69.9% (Andersson, 1991). Lifetime incidences of 'moderate' and 'severe low back pain' of 46.3% (565/1221) and 23.6% (288/1221) respectively were found for American men between eighteen and fifty-five years old (Frymoyer *et al.*, 1983).

The point prevalences of low back pain found for the welders (27%) and CNC workers (16%) are within the range previously reported values by different studies. For the American population, point prevalences of low back pain varied from 12% to 30.2% (Andersson, 1991). At the lower end, a 12.5% point prevalence of low back pain was estimated for Finish men aged 30 or more. The main risks in this sample were occupation (relative risk for herniated disc or sciatica of 4.2 for metal or machine workers), workload, and height (Heliovaara *et al.*, 1987). Similarly, a 12.9% (429/3316) point prevalence of low back pain was found for the Israel population. Here the occupations with highest prevalence were heavy industry workers, bus drivers and nurses (Magora, 1970).

At the higher end, a point prevalence of 23% (95% confidence interval – 95%CI 21-25) was reported for the Swedish population aged between 20 and 59 years. A point prevalence of 26% (95%CI 24-28) was reported for the Australian population, and a 27% (95%CI 26-28) point prevalence was reported for the Dutch population (Picavet and Schouten, 2003; Reigo *et al.*, 1999; Walker *et al.*, 2004). Welders and CNC workers had higher point prevalences of low back pain than reported for Finish men, Israelis, and for some studies of the American population. The welders seem to have higher prevalence than Swedes and comparable to the Netherlands and Australian population, while the prevalence found for the CNC workers was lower than the ones reported for most of the populations. Under-reporting and/or healthy worker effects (only healthy 'strong' workers remain on the job while workers having WLBD move to a lighter job) may partially explain this finding. A possible explanation for the fact that the CNC workers had lower working-life incidence of WLBD and point prevalence of low back pain than the welders but higher annual incidence of WLBD claims, is that the CNC workers were younger, and it may also be due to high turnover in steel company B. There are problems with the epidemiology of WLBD such as classification issues, poor recall, and legal, social and psychological issues/confounders; data is always approximate at best (Andersson, 1991).

7.4.2. Body part discomfort ratings and manual materials handling

The reported weights lifted represented increased risk for the welders and especially for the CNC workers. It may partially explain the higher annual incidence of WLBD claims in CNC workers. It may also explain the trends observed in the discomfort scores where the low back received the highest scores by both groups. The scores given by the welders were higher than by the CNC workers; it may be related to the type of exertion performed where the CNC workers lifted heavier loads for short periods of time while the welders sustained awkward postures with somewhat lighter loads for longer periods of time. These observations are corroborated by the VAS findings and may point toward different mechanisms of discomfort and injury precipitation for the welders and CNC workers. The discomfort reported by the welders may be related to temporary postural syndromes (McKenzie, 1981) while for the CNC workers it may be attributable to overexertion (Kumar, 1994) possibly causing more permanent injuries even though less frequently.

Company A policy stated that 23 kg was the limit for manual lifting, and it was the workers' responsibility to utilize the cranes to lift/move heavier material. Company B policy was not clear; they required workers to use cranes to lift over 27 to 36 kg or over 1.83 m long, and to use flipping bars/chains to flip anything over 90 to 136 kg. The mean values reported were within the companies' limits. The reported weights lifted by welders were also within NIOSH lifting limit (23 kg) for 'optimal conditions' (occasional lifting in the sagittal plane, with good couplings, and vertical displacement of less than 25 cm). It causes 3400 N spinal compression, requires 3.5 Kcal per minute energy expenditure and is acceptable by 99% males and 75% females, but it increases the risk of WLBD moderately (Waters *et al.*, 1993). The reported weights lifted by CNC workers were higher than NIOSH proposed limit. In addition, the lifting conditions for welders and CNC workers were not optimal. Other studies have reported more stringent limits for lifting. Chaffin and Park conducted a one-year longitudinal study about the relationship between WLBD and occupational lifting, including 411 subjects from 103 jobs (Chaffin and Park, 1973). They found increased risk of WLBD when workers lifted more then 16 kg. When the horizontal distance of the load was more than 50 cm from the ankle, lifting loads form 9 to 34 kg represented increased risk of WLBD. Heavy lifting, using jack hammers or machine tools, and motor vehicle operation were demonstrated to be risk factors for 'severe low back pain' in males 18 to 55 years old (Pope *et al.*, 1985).

7.4.3. Rate of Perceived Exertion (Borg scale and VAS)

The Borg scale ratings of perceived exertion are correlated with heart rate (r 0.80-90), minute ventilation and respiratory rate (r 0.76-0.97), and blood lactate during continuous or intermittent exercise on bicycle and treadmill and also during arm or leg work (Borg, 1962, 1982, 1990; Carton and Rhodes, 1985). The exponential (1.6) relationship between physical work and perceived exertion is taken into account by the Borg's 10-point scale where more points in the ratio scale are provided for higher semantic intensity expressions in the categorical correspondent.

In this study, the Borg scale ratings alone (verbal equivalent of strong for both groups) did not differentiate between the welding and CNC jobs. This may be explained by the fact that the Borg scale is a general estimation of the exertion required. The jobs may have different factors contributing to the perceived exertion but still result in similar general perceived exertion. In other words, general indexes are not sensitive to detect specific differences. This hypothesis is supported by the fact that the sum of the VAS scores was also not sensitive to differentiate between the jobs, but the individual VAS scores (scores for each of the five variables studied) pointed out some specific differences between the jobs.

The VAS results are in agreement with the observed job demands and described job tasks. Repetitions and durations are an issue to be considered in both jobs. On the other hand, the postural requirements of the welding job are higher than for the CNC job, while the force requirements are higher in the CNC job in relation to the welding job. These latter two variables allowed differentiating the critical demands of the jobs. This new method of analyzing the physical effort in relation to its different components was useful. Additional research on its use and applications is needed because this method of perceived physical demands analysis is new (Vieira *et al.*, 2006c).

Perceived exertion is related to kinesthetic sensitivity, proprioception, ligament, joint, tendon and muscles cues (mechanoreceptors), and to psychological variables such as personality, motivation and psychometric aptitude. Physical strain of active muscles have higher impact on the perceived exertions at lower levels (work intensity below lactate threshold) such as the levels most often observed during occupational activities, as opposed to high level aerobic exercises where heart rate and blood lactate are better predictors of perceived exertion (Carton and Rhodes, 1985).

7.4.4. Problems and suggestions from the steel workers

The importance of considering the workers' concerns and suggestions was previously demonstrated (Hildebrandt *et al.*, 1996; Udo and Yoshinaga, 2001). The authors of a study of the role of the industrial medical doctor in a steel mill stated that 'the workers themselves could best solve the problem since they knew the working processes best and could, therefore, improve working conditions while taking into account safety and efficiency' (Udo and Yoshinaga, 2001). The workers are rightly concerned about their working postures (Vieira and Kumar, 2004). The risk of WLBD in assembly workers (95 cases and 124 referents) was associated with mild (OR 4.9, 95% CI 1.4-17.4) and severe trunk flexion (OR 5.7, 95% CI 1.6-20.4), and with lateral flexion or rotation (OR 5.9, 95% CI 1.6-21.4) (Punnett *et al.*, 1991). Workers maintaining severe trunk flexion for at least 10% of the working cycle were approximately 9 times more likely to have back disorders than workers maintaining neutral posture (p = 0.003) (Punnett *et al.*, 1991). Asymmetric lifting, as pointed out by the CNC, was demonstrated to be significantly associated with prolapsed lumbar disc (OR 6.1) (Kelsey *et al.*, 1984).

Heavy lifting, use of vibrating equipment, driving, and tobacco consumption were reported to be risk factors for 'severe low-back pain' in men between 18 and 55 years old [lifetime incidence of 23.6% (288/1221)] (Frymoyer *et al.*, 1983). Forty-four percent (163/368) of the subjects with 'no low back pain', 48% (269/565) of the subjects with 'moderate low back pain', and 54% (155/288) of the subjects with 'severe low back pain' lifted 20 kg or more at work. Tobacco consumption, overweight, and sedentary life style were also found related to low back pain in our study (Vieira *et al.*, 2006b).

Overall, the recognized occupational risk factors for WLBD are heavy physical work, static and awkward work postures, prolonged standing and sitting, frequent bend and twisting, lifting, pushing, and pulling, repetitive work, and vibration. The psychological and psychophysical risk factors include work satisfaction and support. Finally, individual risk factors are age and gender, posture, strength, fitness, spinal mobility, smoking, and overweight (AHRE, 2003; Andersson, 1991; Bernard, 1997; BLS, 2001; Chaffin and Park, 1973; Nordin *et al.*, 1997; Picavet and Schouten, 2003; Pope *et al.*, 1985; Punnett *et al.* 1991; Vieira and Kumar, 2004, 2006a; Vieira *et al.*, 2006b).

7.5. Conclusions of Chapter 7

This study presented issues that may lead to high incidences of WLBD in welding and CNC in steel companies. Many established risk factors such as awkward postures, lifting, forceful movements, and heavy workloads were present in the jobs. The questionnaire used was practical and useful to initiate a job evaluation.

No studies using such a combination of measures to evaluate steel workers were found making the present evaluation unique. The workers had good insights into the problems of their jobs and helped identifying risks for WLBD and possible improvements. The information provided by the workers is relevant for assessing and redesigning the jobs, illustrating how the steel workers can significantly contribute to this process. The information presented can be used to design participatory ergonomic interventions aimed at reducing WLBD.

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Chapter 8 – Force and Electromyographic Activity during Pushing, Pulling, and Lifting by Nurses

This chapter presents the results of the force and electromyographic study of pushing, pulling, and lifting by the nurses. A version of this chapter has been submitted for publication in Applied Ergonomics (Vieira and Kumar, 2006b).

8.1. Introduction

Work-related low back disorders (WLBD) are the most prevalent and most costly musculoskeletal disorder (e.g. Woolf and Pfleger, 2003). WLBD is defined as a disorder whose precipitation is related to work demands and are associated with a physical problem causing dysfunction due to tissue damage, affecting the lumbar region of the spine. WLBD nature of injury is classified as ligament sprain, muscle strain, facet joint, and/or sacroiliac joint injury. WLBD occur due to cumulative loading or overexertion in lifting, pushing, pulling, carrying, bending, and/or twisting during work. Thus, WLBD precipitates through the steps of activity, biomechanical stress, temporal loading, tissue strain, and injury (Kumar, 2001; Marras, 2000; McGill, 1997).

Nurses are among the professionals with the highest rates of WLBD (Bejia *et al.*, 2005; Buckle, 1987; Kumar, 2004). Nursing was shown to have odds-ratios for WLBD between 1.8 and 4.3 (Knibbe and Friele, 1996; Lagerstrom *et al.*, 1998; Venning *et al.*, 1987). According to Venning *et al.* (1987), based on a prospective cohort study of 5649 nurses, job related factors are the major predictors of WLBD in nurses. Heavy physical work involving pushing, pulling and lifting in known to significantly increase the risk for WLBD (Bernard, 1997; Frymoyer *et al.*, 1980; Hoogendoorn, 2000).

Studies of WLBD in nurses are common in the literature; however, the problem is still substantial and little success has been achieved in controlling it. Thus, it is important to develop assessment methodologies that allow for designing evidence based ergonomic interventions to control WLBD in nursing jobs.

8.1.1. Background and objectives

Initially, a retrospective epidemiological study and a questionnaire survey were performed (Vieira *et al.*, 2006). Cases of WLBD (n = 159) represented 23% of all injuries, 16% of all first aid, 17% of all medical aid, and 62% of all lost time claim injuries. Seventy-four percent of the WLBD were classified as overexertion injuries (n = 117). Registered and Licensed Practical Nurses, represented 78% (n = 123) of all cases of WLBI recorded in the hospital during the period analyzed and 83% of the lost time claim WLBD cases. Approximately 70% of the WLBD in the Registered and Licensed Practical Nurses happened while transferring or moving patients. The workers with the highest annual incidence of WLBD / hundred workers at the hospital were the orthopedic nurses (ON = 3.7%). The working-time incidence of WLBD and the point prevalence of low back pain were 65% and 30%, respectively (Vieira *et al.*, 2006).

Based on the initial results and on the previous studies available in the literature, the objectives of this study were to test a methodology and to quantify the force and electromyographic activity (EMG) of the erector spinae and rectus abdominis during maximum, job simulated, and preferred levels of pushing, pulling, squat, and stoop lifting by orthopedic nurses. The results were used to define critical aspects of the job requiring intervention to reduce the risk of WLBD; they may also be used to assess the effectiveness of future interventions, as well as to evaluate the functional capacity of new employees and of those returning to work after rehabilitation.

8.2. Methods

This study was approved by the University Research Ethics Committee for Human Studies. Nurses from the orthopedics department of an acute care hospital participated in the study.

8.2.1. Subjects

Twenty-five female orthopedic nurses with no low back disorders resulting in time off work during the previous twelve months participated in the study. On the days of data collection, the clinical educator of the yard was responsible for sending those nurses who had volunteered to participate to a room set aside for data collection.

The volunteer nurses received further explanation about the objectives and procedures of the study, including a statement of their right not to participate and to withdraw at any time with no consequence to them, and signed a consent form. The mean (standard deviation) age of the orthopedic nurses analyzed was 34 (9) years; the height was 167 (7) cm; the body mass was 76 (6) kg, and the body mass index was 26 (5) kg/m^2.

8.2.2. Tasks and levels

The nurses performed randomized pushing, pulling, squat, and stoop lifting tasks. For each task, the following levels of exertions (twelve in total) were performed by the nurses (n = 25): maximum voluntary exertion; job simulated and preferred level which would result in "never going home sore and never getting injured" (Kumar, 1994). The sequence of the twelve exertions was randomized.

The tasks were performed in standardized postures, standing on a board with a mounted poly-system attached to it. The testing apparatus was a poly-system which allowed testing of lifting, pushing and pulling using handles that permitted different grips (Figure 8-1).

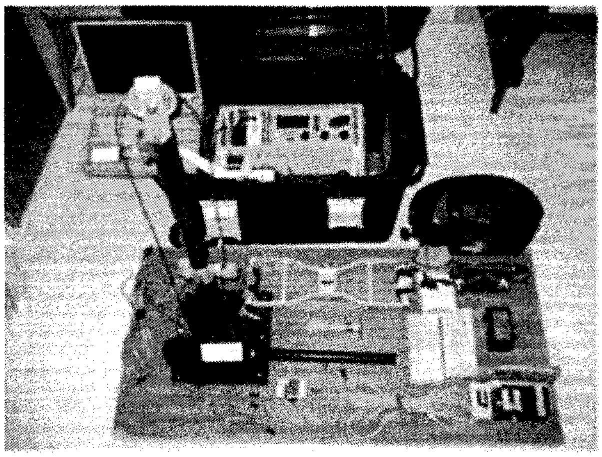

Figure 8-1. Poly-system used for testing pushing, pulling and lifting (stoop and squat).

Pushing and pulling were performed with the nurses standing upright with 90° of elbow flexion. The height of the testing apparatus was adjusted to the nurses' elbow height. Stoop lifting was performed with the knees strait and trunk flexed with the handle positioned at knee height, while squat lifting was performed with the knees flexed at approximately 70° and trunk slightly flexed (approximately 20°), also with the handle positioned at knee height. Figure 8-2 illustrates the testing postures.

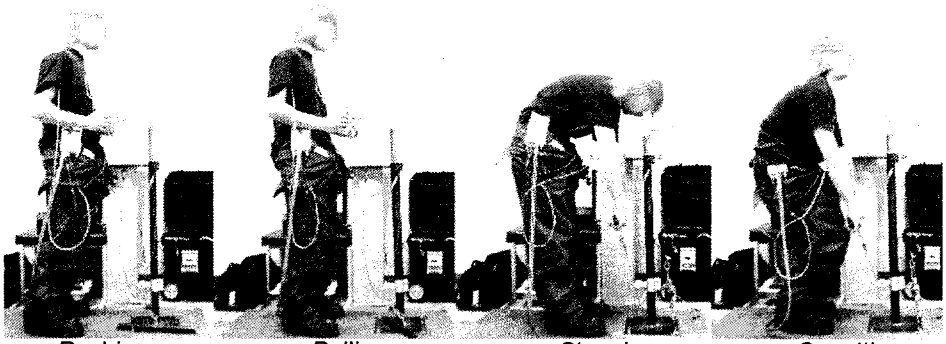

Figure 8-2. Postures during pushing, pulling, stoop and squat lifting.

8.2.3. Equipment used

The equipment used included an EMG device, a load cell, a force monitor, a data link, a metal post and wood base setup, handles, and mobile power supply (battery). The EMG activity was collected using a Delsys Bagnoli 8 electromyography system (Boston, MA, US) including active surface electrodes, electrode cables, preamplifiers and amplifiers. Four pairs of bipolar silver–silver chloride active surface electrodes from Delsys with 10 times onsite amplification and an inter-electrode distance of 1 cm were used. Preamplification at the source, and low impedance active output reduced signal noise. Electrode cables had all conductors protected in order to minimize line interference. The amplifier had AC coupled inputs with a single pole RC high-pass filter with a cut-off frequency of 8 Hz to reject low frequency motion artifacts and obtain all frequencies of surface EMG.

The data acquisition system consisted of an analog-to-digital board with a 100 kHz sampling capacity. The system noise was lower than 5 mV and current leakage lower than 10 mA. The fully-isolated amplifier had additional gain settings up to 10,000 times with a direct current frequency response of 5 kHz and common mode rejection ratio of 92 dB. The synchronized external force generated was recorded using a load cell Omega LCCB-1K (rated output of 3mV/V ± 0.0075mV/V, rated capacity of 1000 pounds). The load cell was fixed to the poly-system.

8.2.4. Experimental procedures

Upon arrival, the subjects were told the purpose and the procedures of the experiment, and signed the informed consent form. The total test session took approximately 40 minutes for each subject. The subjects were familiarized with the procedures. The four pairs of surface electrodes were applied to the subjects after suitable preparation of the skin which was cleaned with an alcohol–acetone mixture (Cram *et al.* 1998). These electrodes were placed using double-adhesive hypo-allergenic tape bilaterally on the bellies of the erector spinae (two pairs of electrodes placed approximately 3 cm to the right and left sides of the midline at L2/L3 vertebrae spinous process level) and rectus abdominis muscles (two pairs of electrodes placed approximately 3 cm from midline and 3 cm above umbilicus). A reference electrode was placed on the acromion. The subjects were randomly asked to exert their maximal, job simulated, and preferred force for a period of 5 s during pushing, pulling, squat and stoop lifting in a randomized order.

Before the maximum voluntary exertion test the subjects were given the following standardized instruction: "When I say go, I want you to bring your force up to your maximum level over 2 seconds and hold for 3 seconds or until I say stop." Similar instructions were given to the subjects before the job simulated tests "bring your force up to your job level". Similarly, in the preferred level tests the subjects were asked to "bring your force up to your preferred level, exert the force you believe you would like to use in your job so that you would never go home sore and never get injured".

8.2.5. Data collection

The surface EMG of the rectus abdominis and erector spinae was recorded bilaterally along with the external force data. The total data collection time for each test was nine seconds. Thus, the data collection started two seconds before initializing force exertion and stopped two seconds after finishing exerting the force to allow recording of the base lines. The verbal instructions during the tests were: "get ready" (at start of recording period), "go" (at 2 seconds), and "stop" (at 7 seconds). The subjects were given one-minute rest period between each test.

The magnitude of the forces in the specific tasks (pushing, pulling, squat and stoop lifting) were quantified, the surface EMG recordings allowed the assessment of the associated muscle load during the activities and the determination of the relative EMG amplitude of the erector spinae and rectus abdominis in the different exertion levels (maximum, preferred, and job simulated). The force (load cell data) and EMG (4 channels) were collected at a sampling rate of 1 kHz in real time. The data were fed to an analogue to digital converter board (Metra Byte DAS 20, Metra Byte Corp., Naunton, MA, US). The converted signals were stored on the hard disc of a Toshiba laptop computer using a DAQ 700 National Instrument data acquisition card.

Data were acquired using modular software. The first module of the software created a subject information file and a random order of the experimental conditions, and it set up these conditions as files for data collection. The second module ran the data sampling and acquisition and recorded the collected data in the previously created files. This phase also converted the force data in Volts into calibrated lbs values based on calibration performed before the experiment. The third module allowed instant graphical display of the collected samples for quality control by displaying all traces and identification of artifacts, with an option to save the trial or re-collect data. The last module of the software permitted a preliminary analysis and calculation of descriptive statistics described next.

8.2.6. Data analysis

The EMG traces obtained during the tests were full-wave rectified and a smoothing routine (7-point linear smoothing repeated once) was used to smooth the signals to reliably interpret the patterns. A sample of approximately three seconds of consistent activity from the five-second exertion was selected by reviewing the processed force and EMG signals. When the torque reached a steady level, a vertical line was drawn marking the start of the steady period of exertion, another line was drawn at the end of the steady exertion providing the time frame (window) for analyzing all channels. The software performed the linear envelope detection of all EMG channels from which it calculated the average EMG of all channels within the selected window.

Since the tasks were performed symmetrically in the sagittal plane and because there were no differences between EMG from the left and right rectus abdominis and erector spinae muscle, the sides were combined in the analysis. The average force and EMG amplitudes during job simulated and preferred levels of exertion were normalized against the subjects' maximal level of exertion during the tasks which was considered as been 100%. The force and EMG activity during the maximum exertion were used to normalize the job simulated and preferred level data for each subject. Thus, the average force and EMG amplitudes measured for both job simulated and preferred exertion levels were normalized against average maximum exertion. Exertion ratios for the job simulated and preferred levels were calculated. The job simulated exertion ratio (JER) is the ratio between the job simulated exertions divided by the maximum (Chaffin and Park, 1973). The preferred exertion ratio (PER) is the job simulated divided by the preferred level of exertion.

8.2.7. Statistical analysis

Statistical analysis was performed using SPSS statistical package (SPSS Inc, Chicago, IL, US) to calculate descriptive statistics including measures of central tendency, measures of variability, percentages, and ratios, and to perform analyses of variance of the average EMG and force. The force, EMG, and exertion ratios between the different levels of exertion and during the different tasks during the different tasks were compared using analysis of variance (one-way ANOVA) with Fisher's least significant difference (LSD) post hoc test. The data distribution was normal and the homogeneity of the variances was checked using Levene statistic. All analyses were based on an alpha-level of 0.05.

The dependent variables (force, EMG, and exertion ratios) were continuous in nature. The analyses were performed individually for the dependent variables because the objective of this study was to determine and compare force and EMG of the erector spinae and rectus abdominis during maximum, job simulated, and preferred levels of pushing, pulling, squat, and stoop lifting. There was no intention of comparing force with EMG or exertion ratios, and no intention of predicting torque from EMG.

The levels of exertion and tasks were compared within themselves only, as opposed to each other. Thus, one dependent variable (force, EMG, or exertion ratios) in three levels of exertion, or in four different tasks were compared. The interactions were of no interest in this study because there was no reason for comparing preferred level of pushing with maximum level of squat lifting, for example.

ANOVAs were used to avoid errors of applying multiple *t*-tests. One-way ANOVAs were used because it is an established way to test the equality of three or more means at one time by using variances. Two-way ANOVAs would have been used if there was interest in evaluating the interactions between levels and tasks, which was not the case in this study. In addition, no repeated measure were collected; the levels were assumed to be independent of each other because the purpose of this study was not to evaluate if the maximum force interfered with the job simulated force, for example. If that were the case, then a repeated measures test would be the most appropriate method (Lomax, 2001). Because this was also not the case, one-way independent ANOVAs were used to compare the levels and to compare the tasks independently.

When significant differences between means were found, a post-hoc test was used to perform multiple pairwise comparisons to identify where the differences were. The LSD test was used because equal variances were determined using Levene statistic and the LSD test allows multiple comparisons assuming equal variances controlling for type I errors (accepting differences when they occur by chance). The LSD test is robust for situations where unequal group sizes are used, which was the case because some levels or tasks had a few more measures than others; it is considered the “most liberal” of the post hoc tests assuming equal variances resulting in lower probability of type II errors (neglect differences when they exist) (Sheskin, 1997).

8.3. Results

8.3.1. Raw Force Data

Figure 8-3 presents the force in Newtons during the tasks at the different levels of exertion (maximum, preferred, and job simulated). The difference between the levels of exertion was significant in all tasks (push, pull, stoop and squat lifting).

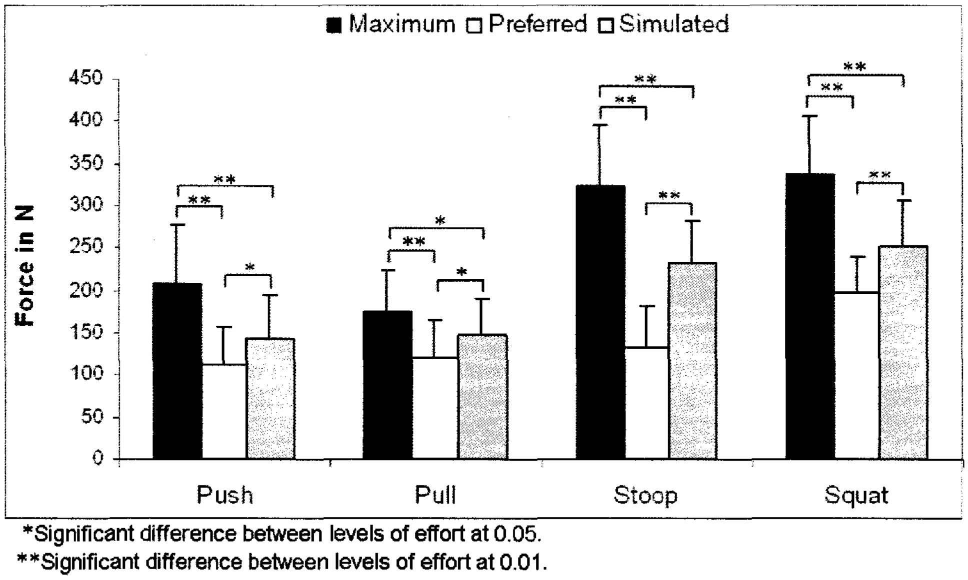

Figure 8-3. Comparison between force levels during push, pull, stoop and squat lifting. The values presented are mean values and the error bars represent one standard deviation.

The comparison of the force during the tasks for each level of exertion showed that during the maximum and job simulated levels of exertion the lifting (squat and stoop) forces were higher than the push and pull forces (mean difference > 84 N, $p < 0.01$). For the preferred level of exertion the force during squat lifting was higher than the forces during the other tasks (mean difference > 63 N, $p < 0.01$).

8.3.2. Maximum EMG

Figure 8-4 presents the average EMG amplitude of the rectus abdominis and erector spinae during maximum push, pull, stoop and squat lifting.

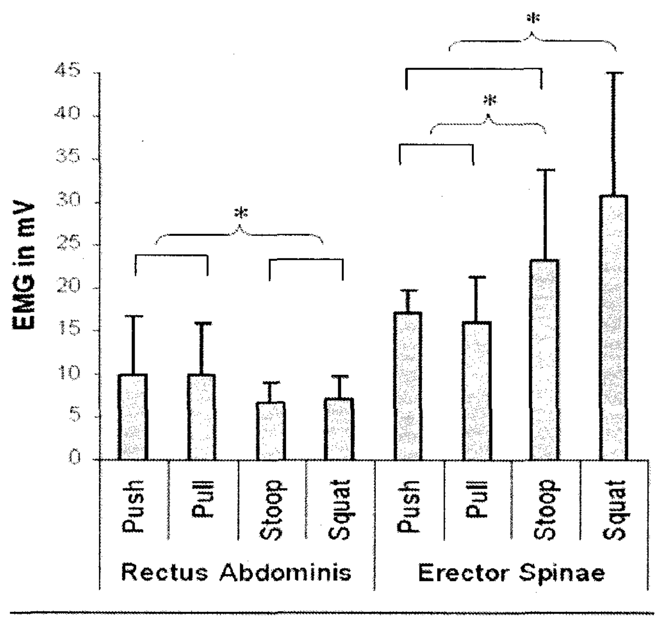

Figure 8-4. Comparison of the EMG of the rectus abdominis and erector spinae during maximum pushing, pulling, stoop and squat lifting. The values presented are mean values and the error bars represent one standard deviation.

The average EMG amplitude of the rectus abdominis during maximum push and pull was higher then during maximum lifting (squat and stoop), while the opposite trend was found for the erector spinae where the average EMG amplitude was higher during maximum lifting than during maximum push and pull; for the erector spinae, the average EMG amplitude during maximum squat was higher than during maximum stoop lifting.

8.3.3. Normalized force and EMG of the erector spine and rectus abdominis during the tasks

In general, the maximum was higher than the job simulated and the preferred levels, and the job simulated level was higher than the preferred level of exertion ($p < 0.001$). Table 8-1 presents the percentage of maximum voluntary exertion for force and EMG of the rectus abdominis and erector spinae during the job simulated and preferred levels of exertion, and the specific significance values for the comparison between job simulated and preferred levels.

Table 8-1. Percentages of maximum voluntary exertion (mean and standard deviation – SD) and significance of the difference (*p*) between preferred and job simulated (Job) levels.

	Force		Rectus Abdominis		Erector Spinae	
Level	Mean	SD	Mean	SD	Mean	SD
Preferred	56	21	69	25	77	32
Job	79	16	82	27	82	29
p	< 0.01		< 0.01		< 0.05	

The percentage of maximum voluntary exertion (% of MVE) for the preferred level was higher for the erector spinae than for the rectus abdominis and force ($p < 0.01$), and the rectus abdominis % of MVE was higher than the force % of MVE (mean difference = 12.72, $p = 0.012$). For the job simulated level, there were no significant differences between the % of MVE for force, for the erector spinae and for the rectus abdominis ($F = 0.6$, $p = 0.549$).

Figure 8-5 presents the percentage of maximum voluntary exertion (mean and SD) for force, rectus and erector / level and task. For force, the difference between preferred and job simulated was significant during all tasks. For the rectus abdominis, the difference was not significant only during pushing, while for the erector spinae the difference was only significant for squat lifting.

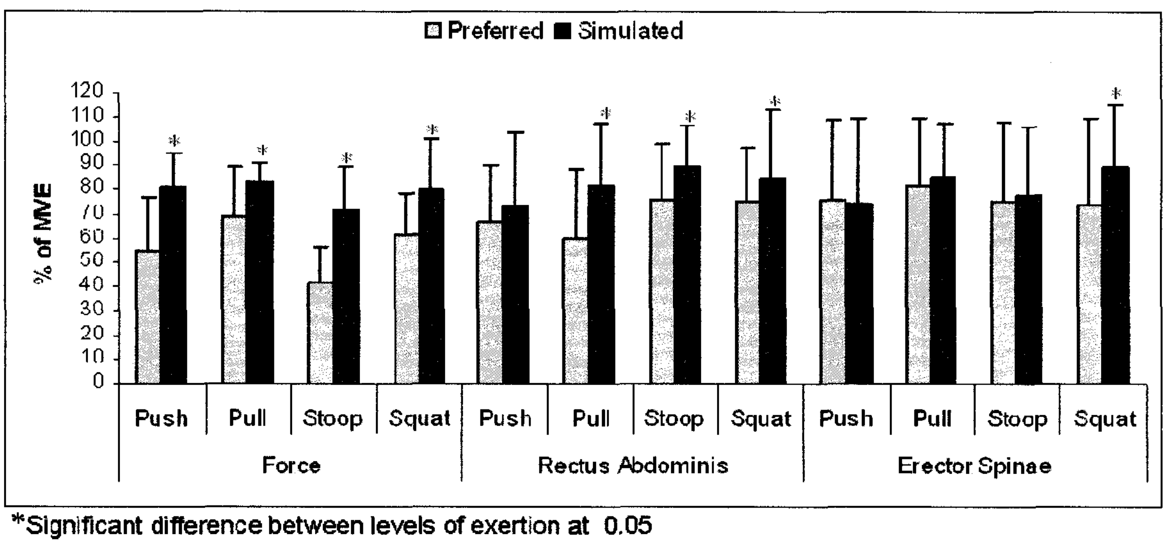

Figure 8-5. Comparison of the percentages of the maximum voluntary exertion (% of MVE) during preferred and job simulated levels for force and EMG of the rectus abdominis and erector spinae during pushing, pulling, stoop and squat lifting.

The maximum exertion was higher than the preferred levels for force and EMG of the rectus abdominis and erector spinae for all tasks ($p < 0.01$). Similarly, the maximum exertion was higher than the job simulated levels in most cases. However, for the EMG of the erector spinae during squat lifting, the difference between maximum exertion and job simulated level was not significant ($p = 0.05$). Table 8-2 presents the mean difference and specific p values for the comparison between preferred and simulated levels of exertion.

Table 8-2. Mean difference (Mean dif.) between preferred (Pref.) and job simulated (Job) levels of exertion and level of significance (p).

Pref. vs. Job		**Push**	**Pull**	**Stoop**	**Squat**
Force	Mean Dif.	-26	-14	-30	-19
	p	<0.001	<0.001	<0.001	<0.001
Rectus Abdominis	Mean Dif.	-7	-22	-13	-9
	p	NS[a]	<0.001	<0.001	0.031
Erector Spinae	Mean Dif.	2	-4	-3	-16
	p	NS	NS	NS	0.003

[a] NS – not significant, $p > 0.05$

8.3.4. Job simulated exertion ratios (JER) and preferred exertion ratios (PER) for force and EMG of the erector spine and rectus abdominis during the tasks

The mismatch between the preferred and job simulated exertion (PER) was the highest for the erector spinae during pushing, and the job simulated exertion was closest to the maximum (JER) for force during stoop lifting. For pushing, the erector spinae had the highest ratio between job simulated and the preferred levels (PER), and the force was closest to the maximum (JER). For pulling, the rectus abdominis had the highest ratios. For stoop lifting, force presented the highest ratios. Finally, for squat lifting, the erector spinae had the highest ratio between preferred and job simulated exertion (PER), and the job simulated activity of the rectus abdominis was closest to the maximum (JER). Table 8-3 presents the exertion ratios for the job simulated and preferred levels.

Table 8-3. Preferred exertion ratio (PER) and job simulated exertion ratio (JER) for force and EMG during pushing, pulling, and lifting (squat and stoop): mean and lower and upper bound of the 95% confidence interval (95%CI).

		Push Mean	95%CI Lower	95%CI Upper	**Pull** Mean	95%CI Lower	95%CI Upper	**Stoop** Mean	95%CI Lower	95%CI Upper	**Squat** Mean	95%CI Lower	95%CI Upper
Force	PER	1.35	1.09	1.60	1.10	0.85	1.34	1.39	1.19	1.60	1.33	1.22	1.44
	JER	0.89	0.83	0.96	0.77	0.62	0.93	0.90	0.78	1.01	0.80	0.71	0.89
Rectus Abdominis	PER	1.11	1.01	1.22	1.19	1.09	1.29	1.12	1.03	1.21	1.57	1.32	1.82
	JER	0.73	0.64	0.82	0.87	0.80	0.94	0.84	0.78	0.91	0.85	0.79	0.91
Erector Spinae	PER	1.75	1.39	2.12	1.12	0.97	1.26	1.27	1.02	1.53	1.68	1.42	1.94
	JER	0.81	0.74	0.89	0.78	0.68	0.87	0.79	0.71	0.87	0.77	0.73	0.82

8.4. Discussion of Chapter 8

The EMG of the erector spinae and the rectus abdominis are directly associated with the resulting forces acting on the spine during the activities recorded (symmetrical lifting, pushing and pulling) (McGill and Norman, 1986). Seroussi and Pope (1987) found that 96% of the variability in the sagittal plane lifting moment could be explained by the sum of erector spinae EMG. As expected we found that there are differences between the maximum, preferred and job simulated levels of force and EMG. The lifting strength was higher than the push and pull strength. The EMG of the rectus abdominis was higher during maximum push and pull than during maximum lifting, while the EMG of the erector spinae was higher during maximum lifting and higher for maximum stoop than maximum squat.

It has been previously found that workers can reliably simulate the forces exerted during the job in common tasks (ICC 0.75-0.95) (Kumar, 1993; van der Beek *et al.*, 1999; Wiktorin *et al.*, 1996). The mean preferred level of lifting force (163 N, SD = 56) was lower than the limit proposed by NIOSH (226 N), while the mean job simulated lifting force (243 N, SD = 52) was higher than the NIOSH lifting limit indicating increased risk of WLBD due to the lifting forces used in the job. Especially because the NIOSH lifting limit is the maximum recommended weight for lifting in "optimal conditions" (occasional lifting in the sagittal plane, with good couplings, and vertical displacement of less than 25 cm), which is not the case in the nursing job. The proposed cut-point is associated with approximately 3400 N of spinal compression, 3.5 Kcal per minute energy expenditure, and is acceptable by 99% males and 75% females, but it increased the risk of WLBD moderately (Waters *et al.*, 1993).

Not surprisingly, the maximum exertion was higher than the preferred for force and EMG of both muscles during all tasks. However, the EMG of the erector spinae during the job simulated squat lifting was not significantly different from its EMG during the maximum squat lifting pointing at an increased risk of injury due to overexertion in squat lifting (McGill, 1997). The job simulated level of exertion represented approximately 80% of the maximum for force and EMG showing the high physical demands of the orthopedic nurses' job.

The job simulated force was higher than the preferred force level for all tasks. Likewise, the job simulated EMG of the rectus abdominis was higher than the preferred level for most tasks, but not for pushing. This may be due to the fact that these muscles are the prime movers in this task (McGill and Norman, 1986). On the other hand, the job simulated EMG of the erector spinae was not significantly different from the preferred level for all tasks but squat lifting. The EMG of the rectus abdominis and especially of the erector spinae during the preferred level of exertion represented high percentages of the maximum voluntary exertion. This finding deserves attention; the preferred level of exertion does not represent the risk free cut-point because the fact that the workers believe it is safe does not mean that no injuries will occur, this may be specially true when considering the cumulative effects of loading the musculoskeletal system (Daynard *et al.*, 2001; Kumar, 1990; Vieira and Kumar, 2006a).

A psychophysical study on maximum acceptable weights (MAW) and isometric strengths during lifting found 23% reduction in the MAW for lifting at 90°, 15% at 60°, and 7% at 30° of asymmetry (Garg and Badger, 1986). For the maximum isometric strength there was 31% decrease at 90°, 21% at 60°, and 12% at 30° of asymmetry. The

MAW was 18 kg plus 53% of the maximum isometric strength. The percentage decrease in maximum isometric strength was higher than the percentage decrease in MAW (Garg and Badger, 1986). This fact further supports the previously discussed issue that the preferred levels may not represent a safe limit for injury precipitation since the percentage of the maximum isometric strength is increasing without being accounted for or noticed by the workers. However, the levels can be used to indicate the mismatch between the preferred level, job simulated exertion, and maximum voluntary exertion, and it may also be used for evaluating the effect of job modifications and training. Normalized instead of absolute strength measures are necessary and require further research (Jaric *et al.*, 2002).

The fact that difference between the preferred and job simulated exertion was the highest for the erector spinae during pushing should be interpreted with caution because the back extensors are not the prime movers during pushing, but act as stabilizers. Thus their activity during the preferred level was low because not much stabilization was required during low levels of contraction of the agonist muscles acting as prime movers (rectus abdominis). On the other hand, the fact that the job simulated force was closest to the maximum during stoop lifting indicates an increased risk of WLBD due to the high lifting demands of the job. Chaffin and Park (1973) reported that high values for the ratio between the force exerted during work and the maximum isometric force of the worker are related to higher incidence of WLBD.

8.5. Conclusions of Chapters 8

The current demands of pushing, pulling and lifting tasks by orthopedic nurses require most of their physical capabilities. The stoop lifting forces of the orthopedic nurses' job are a risk factor for WLBD. This methodology is useful to access the workers functional capacities and compare them with the work physical demands. Job modifications and training programs can be designed and assessed based on these results.

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Chapter 9 – Force and Electromyographic Activity during Pushing, Pulling, and Lifting by Steel Workers

This chapter presents the results of the force and electromyographic study of pushing, pulling, and lifting by the steel workers. A version of this chapter has been submitted for publication in the Journal of Electromyography and Kinesiology (Vieira and Kumar, 2006c).

9.1. Introduction

WLBD is defined as a disorder whose precipitation is related to work demands and is associated with a physical problem causing physical dysfunction due to tissue damage, affecting the lumbar region of the spine. WLBD nature of injury is classified as ligament sprain, muscle strain, facet joint, and/or sacroiliac joint injury. WLBD occurs due to cumulative loading or overexertion in lifting, pushing, pulling, carrying, bending, and/or twisting during work. About 27% of all lost time claims accepted by the Workers' Compensation Board of Alberta (Canada) between 2001 and 2005 were WLBD [Alberta Human Resources and Employment (AHRE), 2006]. The back was the main body part injured and 75% of the WLBD were classified as sprains, strains, and tears; approximately 70% of these resulted from overexertion in lifting, pulling, pushing, carrying, twisting, climbing, tripping, or reaching (AHRE, 2006). Likewise, 65% of all recorded cases of WLBD in 2001 in the U.S. were caused by overexertion, 60% in lifting, 40% in industry laborers, operators, and fabricators (the occupational groups reporting most WLBD) [Bureau of Labor Statistics (BLS), 2001]. Steel workers have a high incidence of WLBD (Dueker *et al.*, 2004; Hildebrandt *et al.*, 1996). About 70% of the workers of the maintenance division of a steel mill had WLBD (Udo and Yoshinaga, 2001). In a study of WLBD in two steel companies (n = 618 workers) the lifetime rate of lumbar symptoms was 66%, and the rate of lumbar symptoms during the previous year and during the previous week were 53% and 25%, respectively (Masset and Malchaire, 1994). Thus, it is important to develop an assessment methodology for evaluating the physical demands and identifying risks to WLBD in order to recommend job modifications and training programs to reduce the risk of WLBD in steel workers.

Heavy physical work involving pushing, pulling and lifting is known to significantly increase the risk for WLBD (Bernard, 1997; Frymoyer *et al.*, 1980; Hoogendoorn *et al.*, 2000). These tasks are frequently performed in the steel industry (International Iron and Steel Institute, 2002). The literature on force and EMG during lifting, pushing and pulling is dated and vast (e.g. Dolan and Adams, 1993; Kroemer, 1974; Kumar *et al.*, 1988; Snook, 1978), but few studies evaluate specifically the functional capacity of steel workers in relation to their physical demands (Marshall and Burnett, 2004). Based on previous estimations, surface EMG has been used in ergonomic studies for more than 50 years (Hagg *et al.*, 2000). A limitation of the previous work in this area is that the studies are most often performed with student samples as opposed to specific groups of workers. In addition, previous studies evaluated the maximum forces and EMG during maximum voluntary contraction (MVC) and specific percentages of MVC (Kumar *et al.*, 2002), or the forces and EMG during work tasks (Marshall and Burnett, 2004; Stalhammar *et al.*, 1986), but most of the studies do not compare the job demands to the workers' capacities (Chaffin and Park, 1973; Vieira and Kumar, 2006a). Previous studies have attempted to determine 'maximum acceptable loads' by having the subjects to perform multiple repetitions of tasks to arrive at the maximal preferred external load (external exposure) (Garg and Badger, 1986; Snook and Ciriello, 1991). However, to our knowledge, no study has attempted to compare the demands to the preferred effort levels by the workers using direct force measures and EMG (internal exposure). This study contributes to the determination of the functional capacity of steel workers and presents an innovative methodology to quantify the workers capacities and compare them to their preferred level and job simulated efforts. It helps to identify the critical aspects of the job requiring intervention to reduce the risk of WLBD.

9.1.1. Background and objectives

Initially, a retrospective epidemiological study and a questionnaire survey of 108 steel workers were performed to assess perceived workload and to identify issues and possible improvements to reduce WLBD (Vieira and Kumar, 2006b). The annual incidence of recorded WLBD was highest for computer numeric control workers (CNC = 5.4%).

The annual incidence of recorded WLBD in the CNC steel workers was higher than the ones reported previously for other heavy jobs involving manual material handling such as for nurses (3.3%), lumbermen (3.3%), and construction workers (2.8%) (Klein *et al.*, 1984). Perceived exertion was strong, and the punching job was considered the most physically demanding and risky job by the steel workers. The participating steel workers perform heavy physical work involving pushing, pulling and lifting (eighteen times per hour on average for each task) (Vieira and Kumar, 2006b). For these reasons, the objectives of this study were to develop and test a methodology and to quantify the force and EMG amplitude of the erector spinae and rectus abdominis during maximum, job simulated, and preferred effort levels for pushing, pulling, stoop and squat lifting by CNC steel workers.

9.2. Methods

This study was approved by the University Research Ethics Committee for Human Studies. Steel workers from the computer numeric control (CNC) sector of a steel blade manufacturing company participated in the study. A more detailed description of the methods is presented on the previous chapter (Chapter 8).

9.2.1. Subjects

Twenty-five CNC steel workers with no low back disorders requiring time off work during the previous twelve month participated in the study. The mean (SD) age of the steel workers analyzed was 27 (5) years; the height was 178 (1) cm; the body mass was 83 (13) kg, and the body mass index was 26 (4) kg/m^2. On the day of data collection, the foreman sent the volunteer workers to a temporary laboratory where they received further explanations about the objectives and procedures of the study, including a statement of their right not to participate and to withdraw at any time with no consequence to them, and signed a consent form. The job of subjects evaluated required to program, set clamps, tighten up and untighten steel blades to machines. The workers loaded and unload the machines manually lifting and positioning the steel blades or using cranes when the blades could not be moved manually. The blades were pushed or pulled on rollers while on machine or tables, which had a coefficient of friction of 0.5. The steel workers aligned steel blades on the conveyor belt so that the punch machine press-stud lined up with the pre-drilled hole, then the machine was activated and the hole was punched in the blade. Figure 9-1 illustrates lifting, pushing and pulling job tasks.

Pushing Pulling Stooping Squatting

Figure 9-1. Tasks performed by the participating steel workers during the job.

The machines punched holes on the blades to fit screws and bolts. Several holes were punched in each blade. The company's lifting policy was not clear and there was no enforcement evident, but it was stated that the workers had to use cranes to lift steel blades over 27 to 36 kg or that were over 1.83 m long. The mean (SD) weight and dimensions of the steel blades were 35 (11) kg, 99 (31) cm long, 25 (5) cm wide, and 3 (0.6) cm thick.

9.2.2. Tasks and levels

The steel workers performed pushing, pulling, squat, and stoop lifting tasks. The following randomized efforts (twelve in total) were performed for each task by each worker (n = 25): maximum effort; job simulated and preferred level which would result in "never going home sore and never getting injured" (Kumar, 1994). The tasks were performed in standardized postures. Pushing and pulling were performed with the steel workers standing upright with 90° of elbow flexion. The height of the testing apparatus was adjusted to the workers' elbow height. The testing apparatus was a poly-system allowing testing lifting, pushing and pulling using handles permitting different grips.

Stoop lifting was performed with the knees strait and trunk flexed with the handle positioned at knee height, while squat lifting was performed with the knees flexed at approximately 70° and trunk slightly flexed (approximately 20°), also with the handle positioned at knee height.

Before the maximum effort test the steel workers were given the following standard instruction: "When I say go, I want you to bring your force gradually up to your maximum level over 2 seconds and hold for another 3 seconds or until I say stop." Similar instructions were given to the steel workers before the job simulated effort tests where the subjects were asked to "bring your force up to your job level". Similarly, in the preferred effort level tests the steel workers were asked to "bring your force gradually up to your preferred level, exert the force you believe you could use in your job so that you would never go home sore and never get injured". During each test, the subjects were asked to exert the force for five seconds. The sequence of the twelve efforts was randomized.

9.2.3. Data collection

The total data collection time for each test was nine seconds. Thus, the data collection started two seconds before initializing the test and stopped two seconds after the test to allow recording the base lines. The verbal instructions during the tests were: "get ready" (at start of recording period), "go" (at 2 seconds), and "stop" (at 7 seconds). The subjects were given one-minute rest period between tests, and the total session took approximately 40 minutes per steel worker.

9.2.4. EMG recording

The magnitude of the forces in the specific tasks (pushing, pulling, squat and stoop lifting) were quantified, the EMG amplitude recordings allowed the assessment of the associated muscle load during the activities and the determination of the EMG amplitude of the erector spinae and rectus abdominis in the different effort levels (maximum, preferred, and job simulated).

A reference electrode (placed on the acromion) and four bipolar, active surface electrodes from Delsys with ten times on-site amplification were attached using double-adhesive hypo-allergenic tape on the bellies of the erector spinae (two electrodes placed approximately 3 cm to the right and left sides of the midline at L2/L3 level) and rectus abdominis muscles (two electrodes placed approximately 3 cm from midline and 3 cm above umbilicus). Skin preparation and electrode placement were performed following Cram *et al.* (1998) guidelines.

9.2.5. Equipment used

The equipment used included an EMG device, a load cell, a force monitor, a data link, a metal post and wood base setup, handles, and mobile power supply (battery). The EMG amplitude was collected using a Delsys Bagnoli 8 electromyography system, at a sampling rate of 1 kHz in real time. The data acquisition system consisted of an analog-to-digital board with a 100 kHz sampling capacity. The fully-isolated amplifier had additional gain settings up to 10,000 times with frequency response DC- 5 kHz and common mode rejection ratio of 92 dB. The synchronized external force generated was recorded using a load cell (Omega LCCB-1K, rated output of 3mV/V ± 0.0075mV/V, and rated capacity of 1000 pounds). The data was recorded to the hard disk of a laptop computer by a data acquisition card. The load cell was fixed to the poly-system.

9.2.6. Data Analysis

The EMG traces obtained during the tests were full-wave rectified, averaged, and linear envelope-detected from the raw EMG signals. From those processed traces, average EMG amplitude was measured. A sample of approximately three seconds of consistent activity from the five-second trial was selected by reviewing the processed EMG amplitude signal of the muscles. Since the tasks were performed symmetrically in the sagittal plane and because there were no differences between EMG amplitude from the left and right rectus abdominis and erector spinae muscle, the sides were combined in the analysis. Descriptive analysis was performed, including measures of central tendency, measures of variability, percentages, and ratios. For each steel worker the job simulated and preferred effort levels of force and EMG were normalized by the maximum effort force and EMG amplitude (normalized = raw/MAX*100). In addition, the force and EMG amplitude from the job simulated efforts were normalized by preferred efforts to evaluate the mismatch between 'required and preferred' efforts; where preferred effort = 100%; thus if job simulated normalized by preferred effort = 150%, job simulated = 1.5 times the preferred effort. The force, EMG amplitude, and ratios during the different tasks were compared using analysis of variance (one-way ANOVA) with Fisher's least significant difference post hoc test. The homogeneity of the variances was checked using Levene statistic. All analyses were based on an alpha-level of 0.05.

9.3. Results

9.3.1. Raw Force Data

Figure 9-2 presents the force in Newtons during the tasks at the different levels of effort. The maximum levels of force were higher than the preferred and job simulated levels during all tasks (push, pull, stoop and squat lifting), but during pull the difference between the preferred and simulated effort levels was not significant.

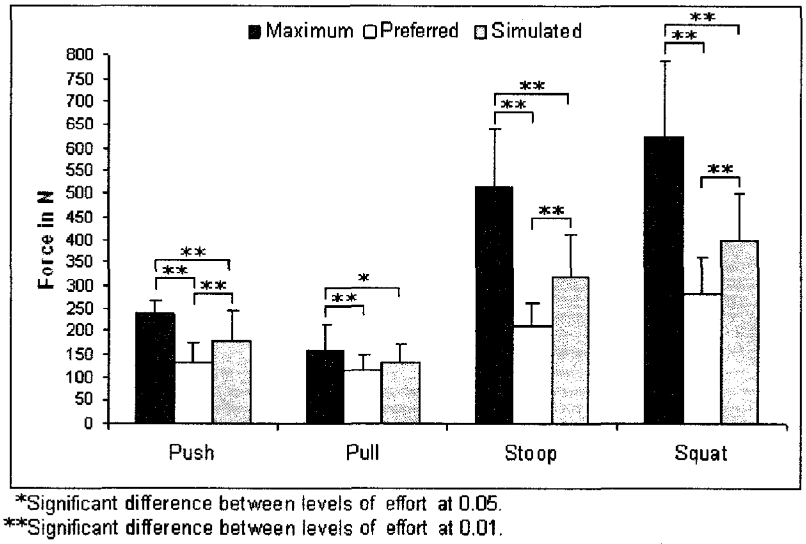


Figure 9-2. Comparison of the forces during pushing, pulling, stooping and squat lifting by steel workers.

The comparison of the force in the different tasks during each level of effort showed similar trends. The squat lifting force was higher than the force during pushing, pulling, and stoop lifting (mean difference > 74 N, $p < 0.01$). The stoop lifting force was higher than the push and pull forces (mean difference > 75 N, $p < 0.01$). The pushing force was higher than the pulling force for maximum and job simulated levels of effort (mean difference > 49 N, $p < 0.03$), but the difference between pushing and pulling forces was not significant for the preferred level of effort (mean difference = 19 N, $p = 0.22$).

The mean (SD) weight of the blades was 343 (108) N, while the job simulated lifting forces were 359 (97) N. The calculated force required to push a 343 N steel blade was 172 N, while the job simulated pushing force was 182 (64) N. Thus, the job simulated forces were reliable.

9.3.2. Maximum EMG Amplitudes

Figure 9-3 presents the amplitude of the EMG of the rectus abdominis and erector spinae during maximum pushing, pulling, stoop and squat lifting.

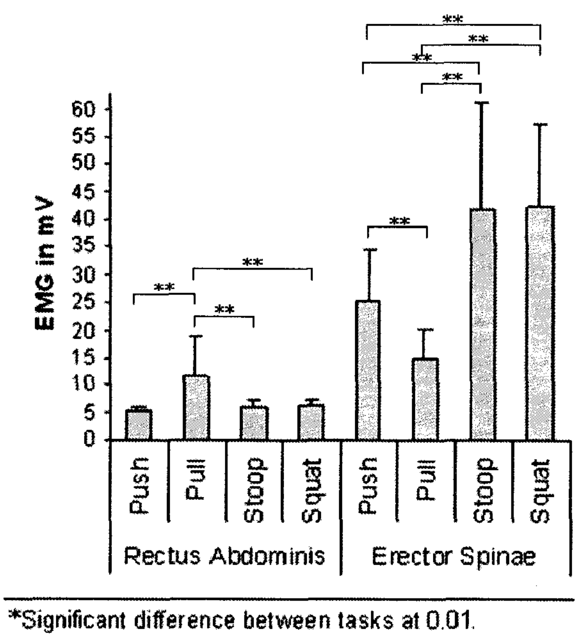


Figure 9-3. EMG amplitude of the rectus abdominis and erector spinae during maximal push, pull, stoop and squat lifting by steel workers: comparison between tasks.

The amplitude of the EMG of the rectus abdominis for maximum pulling was higher than during the other tasks (maximum pushing, squat and stoop lifting). The EMG amplitude of the erector spinae was higher during maximum lifting (stoop and squat) than during maximum pushing and pulling; no significant differences were found between maximum squat and stoop lifting, but the EMG amplitude of the erector spinae was higher in maximum pushing than in pulling.

9.3.3. Normalized Job Simulated and Preferred Efforts for Force and EMG Amplitude

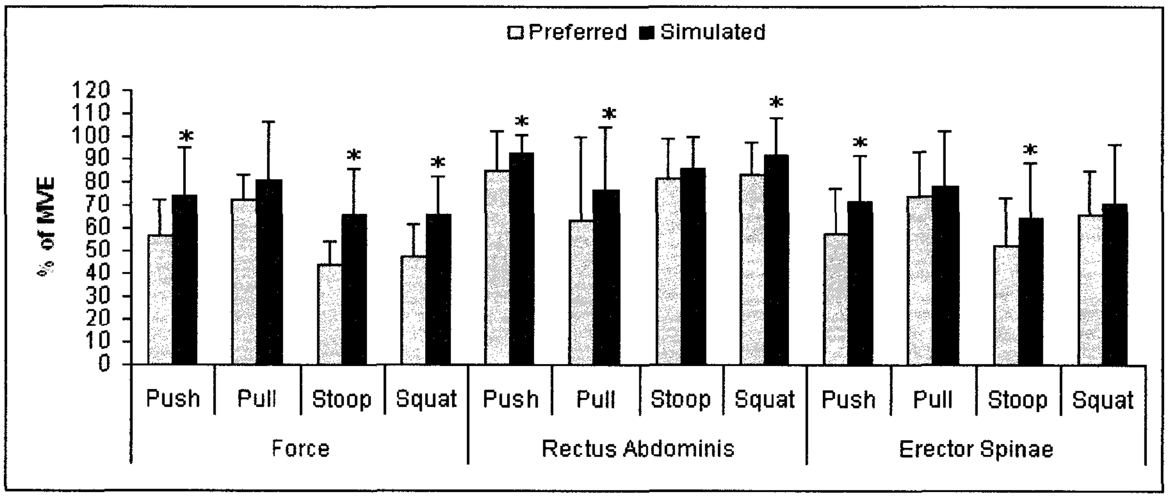
In general, the maximum effort (ME) was higher than the job simulated and the preferred effort levels, and the job simulated effort was higher than the preferred effort level ($p < 0.01$). Table 9-1 presents the percentage of maximum effort (% ME) for force and EMG amplitude of the rectus abdominis and erector spinae during the job simulated and preferred levels of effort, and the specific significance values for the comparison between job simulated and preferred effort levels.

Table 9-1. Percentages of maximum effort (mean and standard deviation – SD) and significance of the difference (p).

	Force		**Rectus Abdominis**		**Erector Spinae**	
Level	Mean	SD	Mean	SD	Mean	SD
Preferred	55	17	78	25	62	22
Simulated	72	22	87	19	71	24
p	< 0.001		< 0.001		< 0.001	

For the preferred level, the % ME was higher for the rectus abdominis than for the erector spinae and force (mean differences = 16% and 23%, respectively, $p < 0.01$ for both comparisons), and it was higher for the erector spinae in relation to force (mean difference = 7%, $p = 0.012$). For the job simulated level, the % ME for the rectus abdominis was also higher than for the erector spinae and force (mean differences = 15% and $p < 0.01$ for both comparisons), but there were no differences between the force and the erector spinae % ME (mean difference = 0.46, $p = 0.866$).

The maximum effort was higher than both preferred and job simulated levels for force and EMG amplitude of the rectus abdominis and erector spinae for all tasks ($p < 0.01$). Figure 9-4 presents the percentage of maximum effort (mean and SD) for force and EMG amplitude of the rectus abdominis and erector spinae per level of effort per task.



*Significant difference between preferred and job simulated levels at 0.05

Figure 9-4. Comparison of the percentages of the maximum effort (% ME) during preferred and job simulated levels for force and EMG amplitude of the rectus abdominis and erector spinae during pushing, pulling, stoop and squat lifting.

The normalized job simulated force was higher than the preferred level for most tasks, but not for pulling. The normalized EMG amplitude of the rectus abdominis in the job simulated stoop lifting effort was not different from its normalized EMG amplitude in the preferred stoop lifting effort. The normalized job simulated EMG amplitude of the erector spinae was significantly higher than the preferred level for pushing and stooping, but not for pulling and squatting.

9.3.4. Mismatch between Job Simulated and Preferred Efforts for Force and EMG Amplitude

In general, the job simulated force represented 127 (53) % of the preferred force level. The EMG amplitude of the erector spinae during the job simulated effort represented 132 (52) % of the EMG amplitude during the preferred effort, and the EMG amplitude of the rectus abdominis during the job simulated effort represented 117 (31) % of their EMG amplitude during the preferred effort. The mismatches between the preferred and job simulated efforts for force and amplitudes of the erector spinae and rectus abdominis' EMGs were not significantly different. On the other hand, the distance between the preferred and job simulated efforts was higher for the EMG amplitude of the erector spinae in relation to the EMG amplitude of the rectus abdominis (mean difference

= 15%, p = 0.001). Table 9-2 presents the job simulated efforts normalized by the preferred efforts (preferred effort = 100%) for force and amplitudes of the erector spinae and rectus abdominis' EMGs during each task (push, pull, stoop and squat lift).

Table 9-2. Job simulated efforts expressed as percentages of the preferred efforts for force and EMG amplitude during pushing, pulling, squat and stoop lifting: mean, lower and upper bounds of the 95% confidence interval (95%CI).

	Push Mean	95%CI Lower Upper	**Pull** Mean	95%CI Lower Upper	**Stoop** Mean	95%CI Lower Upper	**Squat** Mean	95%CI Lower Upper
Force	106	98 115	145	120 171	110	86 135	148	129 168
Rectus Abdominis	113	106 120	112	105 119	114	103 124	129	119 139
Erector Spinae	150	129 172	118	108 128	128	115 141	132	119 145

The mismatch between job simulated and preferred efforts was highest for the erector spinae during pushing (mean difference > 36%, $p < 0.002$). For pulling, force had the highest mismatch between preferred and job simulated efforts (mean difference > 26%, $p < 0.006$). For stoop lifting, the erector spinae had higher mismatch and, for squat lifting, force had the higher mismatch between preferred and job simulated effort, but the differences between force and EGM amplitude of both muscles were not significant for lifting.

The comparisons within force and muscles between tasks showed that, for force, the mismatch between preferred and job simulated efforts was higher during squat lifting and pulling than during stoop lifting and pushing (mean difference > 34%, $p < 0.016$). For the rectus abdominis, the mismatch was the highest during squat lifting (mean difference > 15%, $p < 0.015$), and for the erector spinae the percentage of the preferred effort level used was higher during pushing in relation to pulling and stooping (mean difference > 21%, $p < 0.04$).

9.4. Discussion of Chapter 9

The amplitude of the EMGs of the erector spinae and rectus abdominis are directly associated with the resulting forces acting on the spine during symmetrical lifting, pushing and pulling (McGill and Norman, 1986). The lifting strength (maximum voluntary force) was higher then the pushing and pulling strengths. As anticipated, there were differences between the maximum, preferred and job simulated effort levels for force and EMG amplitudes. The job simulated forces were higher than the preferred force level for pushing, stoop and squat lifting. However, the difference between preferred force and job simulated force was not significant for pulling which is not a very demanding task in the job of the steel workers analyzed in this study.

Our results are consistent with the findings of previous studies that have found that workers can reliably simulate the forces exerted during the job in common tasks (ICC = 0.75-0.95) (Kumar, 1993; van der Beek *et al.*, 1999; Wiktorin *et al.*, 1996). The preferred lifting force (247 N, SD = 75) was similar to the limit (226 N) proposed by the National Institute for Occupational Safety and Health (NIOSH), while the job simulated lifting force (359 N, SD = 104) was higher than the NIOSH lifting limit. This indicates an increased risk of WLBD due to the lifting forces used in the job, especially because the NIOSH lifting limit is the maximum recommended weight for lifting in "optimal conditions" (occasional lifting in the sagittal plane, with good couplings, and vertical displacement of less than 25 cm), which is not the case in the job of the steel workers analyzed. The proposed cut-point is associated with approximately 3400 N of spinal compression, 3.5 Kcal / minute energy expenditure, and is acceptable by 99% males and 75% females, but it increased the risk of WLBD moderately (Waters *et al.*, 1993).

The EMG amplitude of the rectus abdominis during pulling was higher then during pushing, squat and stoop lifting. This fact is interesting because during pulling this muscle acts as a trunk stabilizer as opposed to a prime mover. As expected, the EMG amplitude of the erector spinae was higher during lifting than during pushing and pulling, but similarly to the behavior of the rectus abdominis, it was higher for pushing than for pulling, where in pushing the erector spinae has a trunk stabilization role (McGill and

Norman, 1986). Not surprisingly, the maximum effort was higher that the preferred effort level for force and EMG amplitude of both muscles during all tasks.

The mean job simulated effort represented between 70% and 80% of the maximum for force and EMG amplitudes showing the high physical demands of the steel workers' job. However, the normalized EMG findings should be interpreted knowing that the maximum effort EMGs were the muscles activities during the maximum effort tests for lifting, pushing and pulling, as opposed to the maximum voluntary contraction (MVC) EMGs during tests designed to recruit the specific muscles. The EMG of the muscles during MVC would likely be higher than the EMG of the muscles during the maximum effort (ME) in the tasks evaluated. This clarification helps to explain why the percentages of ME were higher for the EMG than for force.

Since the rectus abdominis are not strongly recruited in the tasks analyzed, their % ME for the preferred and job simulated levels were higher than for force and EMG of the erector spinae. Similarly, since we collected the erector spinae activity during the tasks as opposed to during their specific MVC, their % ME were higher than for force during the preferred effort. However, the erector spinae % ME was closer to the force % ME because these muscles are more active during the tasks studied. Actually, there were no significant differences between the erector spinae EMG amplitude and force percentages of ME for the job simulated level. This may be understood by the fact that the sum of erector spinae EMG amplitudes explained 96% of the variability in the sagittal plane lifting moment (Seroussi and Pope, 1987). Thus, the erector spinae EMG amplitude during the lifting tasks was probably closer to their specific MVC.

The EMG amplitude of the rectus abdominis during the job simulated effort was higher than during the preferred effort for most tasks, but not for stooping when this muscle in not very active (McGill and Norman, 1986). The EMG amplitude of the erector spinae during the job simulated effort was only higher than the preferred level for pushing and stoop lifting. The EMG amplitude of the erector spinae and especially of the rectus abdominis during the preferred level of effort represented high percentages of the maximum effort. This finding deserves attention; the preferred level of effort does not represent the risk free cut-point because the fact that the workers believe it is safe does not mean that no injuries will occur, this may be specially true when considering the cumulative effects of loading the musculoskeletal system (Kumar, 1990; Norman *et al.*, 1998; Vieira and Kumar, 2006a).

A psychophysical study on acceptable weights and isometric strengths during symmetric and asymmetric lifting found 23% reduction in the maximum acceptable weights for lifting at 90°, 15% at 60°, and 7% at 30° of asymmetry (Garg and Badger, 1986). In relation to the maximum isometric strength, there was 31% decrease at 90°, 21% at 60°, and 12% at 30° of asymmetry. The maximum acceptable weight was 18 kg plus 53% of the maximum isometric strength. The percentage decrease in maximum isometric strength was higher than the percentage decrease in maximum acceptable weight (Garg and Badger, 1986). This fact further supports the previously discussed issue that the preferred levels may not represent a safe limit for injury precipitation since the percentage of the maximum isometric strength is increasing without being accounted for or noticed by the workers. However in the absence of any better tool, the levels can be used to indicate the mismatch between the preferred level, job simulated effort, and maximum effort, and it may also be used for evaluating the effect of job modifications and training programs.

Normalized instead of absolute strength measures are necessary and require further research (Jaric *et al.*, 2002). The approach used to capture the preferred level is an innovative aspect of the methodology used and was based on the theoretical framework proposed by Kumar (1994). This study is the first step towards its validation. The stability of the preferred force level measures [CV = 28 (3) %] was comparable to the stability of the job simulated [CV = 30 (4) %] and maximum force measures [CV = 24 (9) %]. These results demonstrate the inter-subject reliability of this measure. Future studies could evaluate the intra-subject reliability (repeated measures) and the predictive validity of the preferred levels (if a job is designed according to the workers preferred levels of force, does it result in decreased rates of injury?).

The job simulated force effort during pushing was closest to the maximum effort. In addition, pushing caused the highest mismatch for the erector spinae activity between the preferred and job simulated efforts, and the activity of the rectus abdominis was closest to the maximum also during this task. High values for the ratio between the force exerted during work and the maximum isometric force of the worker are related to higher incidence of WLBD (Chaffin and Park, 1973). Thus, based on the results, the pushing forces are the critical aspect of the punching job of the computer numeric control steel workers.

9.5. Conclusion of Chapter 9

The current physical demands during pushing and lifting job exertions by the computer numeric control steel workers require most of their physical capabilities. The lifting and pushing forces of the job of steel workers analyzed are risk factors for work-related low back disorders (WLBD). Based on the results, pushing is the critical task requiring intervention. This methodology is useful to access the workers functional capacities and compare them with the work physical demands. Job modifications and training programs can be designed and assessed based on these results. Further studies should evaluate if job modifications and/or training programs designed based on the information gathered using this methodology can successfully reduce the incidence of WLBD.

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Chapter 10 – Comparison of Force and Electromyographic Activity during Lifting, Pushing and Pulling By Nurses and Steel Workers

This Chapter presents an abstract presenting a comparison of the force and EMG of the nurses and steel workers. This abstract was presented and published in the Proceedings of the XVI Congress of the International Society of Electrophysiology and Kinesiology (Vieira and Kumar, 2006).

10.1. Aims

The objectives of this study were to evaluate the maximum, job required, and preferred level of force and electromyographic activity (EMG) of nurses and steel workers during squat and stoop lifting, pushing and pulling.

10.2. Methods

Ten male steel workers and ten female nurses participated in the study. The force and EMG of the erector spinae (bilateral electrodes placed 3 cm of the midline at L2/L3 level) and rectus abdominis (bilateral electrodes placed 3 cm from midline and 3 cm above umbilicus) were recorded. Three five-second trials during squat and stoop lifting, pushing and pulling were done: maximum voluntary force (MVF), job required, and preferred working level to "never go home sore and never get injured" (12 tests/subject).

10.3. Results

The steel workers and nurses' MVF were respectively 579±128 N and 314±88 N for squat lifting ($p<0.002$); 559±186 N and 314±88 kg for stoop lifting ($p<0.002$); 255±88 N and 216±78 N for pushing ($p>0.05$), and 157±49 kg and 186±78 N for pulling ($p>0.05$). The MVF during lifting was higher than during push and pull for both groups ($p<0.02$). For the steel workers, the erector spinae EMG during lifting was higher than during push and pull ($p<0.046$). Push tended to be higher than pull; this was significant for the left erector spinae EMG ($p=0.007$). The opposite relation was found for the rectus abdominis EMG, which tended to be higher during pull; this was significant in relation to push ($p<0.024$).

For the nurses, the EMG of both muscles was higher during squat lifting ($p<0.02$). This finding may be explained by an increased handle distance during squat lifting because it did not fit between the knees. Figure 10-1 presents the required and preferred force level (% of MVF) during squat and stoop lifting, pushing and pulling.

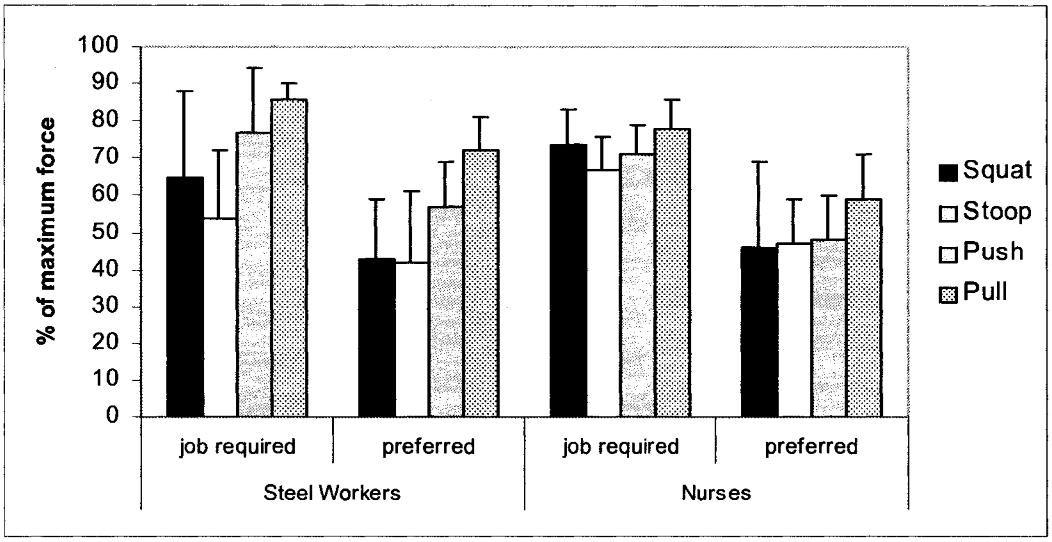

Figure 10-1. Job required and preferred force level as a percentage of the maximum voluntary force of the steel workers and nurses during squat lifting, stoop lifting, pushing, and pulling.

The job required exertion tended to be higher than the preferred level for both groups. Only for stoop lifting by the steel workers the difference between job required and preferred force level was not statistically significant. This may be explained by the fact that these workers tend not to perform this type of lifting during the job. Only the preferred pull level was higher for the steel workers than for the nurses ($p=0.015$). Only for pull the difference between the right erector spinae EMG during the job required exertion and during the preferred level was not statistically significant. For the left erector spinae, this difference was only significant for squat lifting. This shows that even in symmetrical tasks there are differences between the left and right erector spinae muscles. No significant differences were found for the rectus abdominis EMG.

10.4. Conclusions of Chapter 10

There are workload differences between the jobs and gender differences in physical capacity. This methodology is useful to identify problems/risks in jobs with a high incidence of musculoskeletal disorders. Job modifications and training programs can be designed and assessed based on these results. Further studies could evaluate if the designed interventions based on the information gathered using this methodology can successfully reduce the incidence of work-related musculoskeletal disorders.

Reference of Chapter 10

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Chapter 11 – Motion and Biomechanical Demands of Patient Handling Tasks

This chapter presents the results of the study of the range of motion, motion during the job and biomechanical loads on the lumbar spine of nurses during patient handling and transfers. Thirty-six volunteer female nurses participated in the cross-sectional study of 9 nursing tasks. The lumbar range of motion (ROM) and motion during nursing the tasks were measured. The compression and shear forces at L5/S1, ligament strain and percent population without sufficient torso strength to perform 14 phases of the nursing tasks were estimated. A version of this chapter has been submitted for publication in Nursing Research (Vieira and Kumar, 2006c).

11.1. Introduction

Precipitation of work-related low back disorders (WLBD) is related to biomechanical demands and is associated with a physical problem causing physical dysfunction due to tissue damage, affecting the lumbar region of the spine (Kumar, 2001; Marras, 2000; McGill, 1997). WLBD in nurses occurs due to cumulative loading or overexertion in lifting, pushing, pulling, carrying, bending, and/or twisting during work (Brulin *et al.*, 1998; Lagerstrom *et al.*, 1998; Smedley *et al.*, 1995). WLBD are the most prevalent and most costly work-related musculoskeletal disorder (e.g. Woolf and Pfleger, 2003). The prevalence of low back pain in nurses is even higher than in the general population (e.g. Josephson *et al.*, 1997; Lagerstrom *et al.*, 1998). Nurses are among the professionals with the highest rates of WLBD (Bejia *et al.*, 2005; Buckle, 1987; Kumar, 2004). Nurses have the highest incidence of disabling WLBD among all professionals in the US (Jensen, 1987). Similarly, the rates of WLBD in nurses are also high in Australia, Brazil, Canada, China, England, France, Italy, Japan, Korea, Sweden, and other countries (e.g. Ando *et al.* 2000; Gurgueira *et al.* 2003; Lagerstrom *et al.*, 1995; Larese and Fiorito, 1994; Niedhammer *et al.*, 1994; Smith *et al.*, 2004 and 2005a,b; Vieira *et al.*, 2006). For example, Engkvist *et al.* (1992) found that female nursing aides had WLBD six times more often than all other female workers in Sweden; lifting was the cause of 84% of the WLBD.

Awkward postures, heavy physical work, lifting, forceful movements and whole body vibration have causal relationship with WLBD (Bernard, 1997; Frymoyer *et al.*, 1980). A prospective cohort study found that job related factors were the major predictors of WLBD in 5649 nurses (Venning *et al.*, 1987). Manual patient transfers were previously reported by nurses as the most stressful transfer method and these tasks were previously reported as the main cause of WLBD among nurses (Garg and Owen, 1994; Morlock *et al.*, 2000; Yassi *et al.*, 1995). Hignett (1996) reported that the greater the manual handling of patients, the higher the incidence rate of WLBD. Stobbe *et al.* (1988) compared the incidence of WLBD in two groups of nurses; one group was classified as having high frequency of manual patient transfers (more than five manual transfers per shift, n = 317) and the other group was classified as having low frequency of manual patient transfers (less than two manual transfers per shift, n = 98). The low transfer frequency group "survived" longer without a WLBD than the high frequency group (10% difference after 1215 days, $p < 0.01$). The licensed practical nurses in the high manual transfer frequency group were 7.54 times more likely to have a WLBD than those in the low frequency group (Stobbe *et al.*, 1988).

It is important to quantify the biomechanical demands of the nursing tasks to understand their relationship with the high rates of WLBD and to develop and access control programs and interventions. A generally accepted approach is to identify tasks with high risk, determine the critical factors in the tasks, and modify those critical factors to reduce the risk of WLBD. Studies of WLBD in nurses are common in the literature; however, the problem is still substantial. For these reasons, it is important to quantify the demands to design evidence based interventions to control WLBD in nursing jobs.

11.1.1. Background and objectives

A retrospective epidemiological study (review of five-year injury records) and a questionnaire survey (47 nurses) were performed (Vieira *et al.*, 2006). WLBD (n = 159) represented 23% of all injuries, 16% of first aid injuries, 17% of medical aid injuries, and 62% of lost time injuries in the hospital evaluated. Seventy-four percent of the WLBD were overexertion injuries; 78% and 83% of all WLBD resulting in time off work occurred among nurses.

Approximately 70% of the WLBD in nurses happened while transferring or handling patients in bed and the departments where WLBD occurred most often in the entire hospital were orthopaedics (32%) and ICU (17%). The working-life incidence of WLBD was 65% among orthopedic nurses (ON) and 58% among ICU nurses (IN). The point prevalence of low back pain was 30% in the ON and 25% in the IN. The mean (SD) weight handled was reported to be 47 (30) kg by the ON and 26 (10) kg by the IN. The rate of perceived job exertion on Borg's 10-point scale was 7 (2) or very strong for the ON, and 6 (2) or strong for the IN. Patient transfers, turning and repositioning patients in bed were considered the most physically demanding tasks of the job by the ON and IN, respectively (Vieira *et al.*, 2006).

A study of the forces and EMG of the rectus abdominis and erector spinae during maximum voluntary exertion (MVE), job simulated, and preferred levels for lifting (squat and stoop), pushing, and pulling was performed, and included 25 nurses (Vieira and Kumar, 2006a). The job simulated force [79 (16) % of MVE] was higher than the preferred level [56 (21) % of MVE, $p < 0.01$]. Based on the results of the initial studies, it was decided to quantify the biomechanical demands of patient transfers in the orthopedics department (ON tasks) and the biomechanical demands of turning and repositioning patients in bed in the ICU (IN tasks). Thus, the objectives of this study were to advance the knowledge in nursing by (I) quantifying the biomechanical demands of manual patient transfers by orthopedic nurses (ON) and of turning and repositioning patients in bed by intensive care nurses (IN), and by (II) providing evidence based recommendations to reduce the risk of low back disorders.

11.2. Materials and Methods

11.2.1. Subjects

Female nurses from the orthopedics department (21 ON) and from the ICU (15 IN) of an acute care hospital participated in the study. The inclusion criterion was to have no WLBD resulting in time off work during the previous twelve month. The mean (SD) ages for the ON and IN were 35 (7) and 34 (9) years, weight was 74 (8) and 68 (6) kg, height was 168 (5) and 167 (7) cm and body mass indices were 26 (5) and 25 (4) kg/m^2, respectively.

Before data collection, the subjects were informed about the study, and signed the consent form. This study was approved by the University and Hospital Research Ethics Committee for Human Studies.

11.2.2. Lumbar range of motion

In order to normalize the data and allow the calculation of the thresholds for analysis, the maximum lumbar range of motion (ROM) of the nurses was assessed. Three maximum flexion, extension, lateral flexion (right and left), and rotation (right and left) trials were performed in each direction. The sequence of the trials and the directions were randomized. The nurses were instructed to "bend forward/backward as far as you can", "bend to the right/left side as far as you can" and to "rotate to the right/left side as far as you can".

The lumbar ROM on the frontal and transversal planes was measured using electrogoniometers (elgons) including one goniometer XB 180 and one torsiometer Q1 50 (Biometrics, Gwent, UK). The lumbar spine was palpated and the spinal processes of T12 and L5 were marked. The elgons were placed in attachment ducts and were attached to the skin over the landmarks using double-adhesive tape, and the connection cables were attached to the elgons and respective channels on the data link (Vieira and Coury, 2002). The data was transferred to a laptop computer. The operating temperature of the elgons is from 0° to 40°C; the operating humidity range is from 30% to 75%; the repeatability error is $\leq 1°$, and the maximum measurement error is ± 3° and ± 5° when measuring single and multiple planes (Biometrics Ltd., 1999). The small repeatability and measurement errors of the back elgons were confirmed by independent studies (Shiratsu and Coury, 2003; Vieira and Coury, 2002).

The elgons were not used for sagittal ROM measures due to the potentially hazardous stretching of the devices. The extreme flexion ranges could overstretch the wire that keeps the endblocks of the elgons together and contains the strain gauges that measure the angular displacement (Biometrics Ltd., 1999). The lumbar ROM in the sagittal plane was measured using perpendicular markers photogrammetry (Vieira and Coury, 2004). This method has been used to measure the lumbar posture and sagittal motion at least since 1974 (Kumar, 1974).

The concurrent validity of this method to measure the orientation of lumbar segments was found to be good when comparing to X-ray measurements (r^2 from 0.91 to 0.98) (Chen and Lee, 1997). The parallel reliability of sagittal plane measures of lumbar motion using perpendicular markers photogrammetry was also found to be high when compared with elgons' measures [r^2 = 0.994; measurement error: 0.6° (0.7)] (Vieira and Coury, 2004).

The perpendicular markers were placed on the skin over the landmarks using double-adhesive tape and single- adhesive tape was put on top. The nurses were video-taped from the sagittal plane (left side) to measure the flexion and extension ROM. Thirteen frames were extracted from video using windows movie maker to represent the neutral posture (7 frames), fully flexed posture (3 frames), and fully extended postures (3 frames). An audio video interleave (AVI) movie was created with the 13 jpeg pictures using jpegAvi software. The new avi video was resaved using the VirtualDub software to allow for analysis using the PostureAna software (University of Alberta). The origin and tips of the perpendicular markers were marked clockwise on the 13 frames and the angles were calculated. The difference between the neutral posture and the fully flexed posture defined the flexion ROM and the difference between the neutral and fully extended postures defined the extension ROM. Figure 11-1 illustrates the measure of lumbar ROM in the sagittal plane using perpendicular markers photogrammetry.

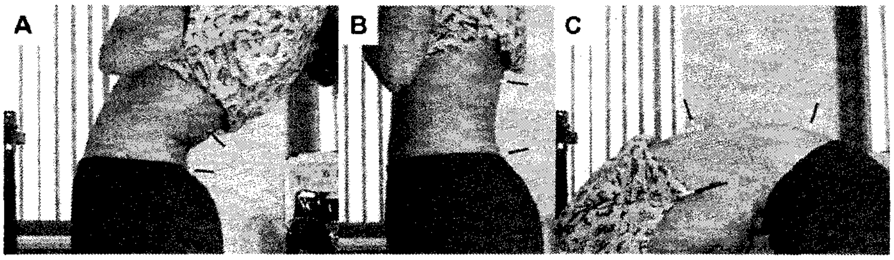

Figure 11-1. Measure of lumbar range of motion in the sagittal plane using perpendicular markers photogrammetry: A = extension; B = neutral; C = flexion.

11.2.3. Lumbar motion during the nursing tasks

The lumbar motion in the sagittal, frontal, and transversal planes during patient transfers from stretcher to bed, bed to stretcher, bed to chair, chair to bed, chair to wheelchair, wheelchair to chair, turning patients towards and away, and repositioning patients up in bed was recorded using the elgons.

The elgons were fixed leveled with T12 and L5. The nurses performed the tasks and the data was transferred to a computer using the Biometrics Data Link. The lumbar motion on the frontal plane [left (+) and right (-) lateral flexion] was recorded on Channel 1; the lumbar motion on the sagittal plane [flexion (+) and extension (-)] was recorded on Channel 2, and the lumbar motion on the transversal plane [left (+) and right (-) rotation] was recorded on Channel 3.

The peak and average lumbar motion during the tasks were determined using the elgons' software (Biometrics, Gwent, UK). In addition, the tasks were video-taped from the sagittal plane at 30 Hz using a Canon mini DV media digital camera. An event marker was recorded on Channel 8 of the elgons, a switch activated a red light LED which appeared in the corner of the foreground of the film and generated a spike on Channel 8 of the data logger. Thus, the video-recordings and elgons data were synchronized by means of a LED and an electric pulse (step increase in voltage) recorded along with the elgons' data.

11.2.4. Lumbar compression and shear forces, ligament strain and population without sufficient torso strength to perform the job: procedures and load estimation

The weight and height of the nurses were measured. The videos were used to measure the postures and to determine beginning, end, and duration of the tasks and their different phases. Round surface markers covered with highly reflective tape were placed on the ankle, knee, hip, shoulder, elbow, and wrist regions to facilitate joint location identification. The phases of the tasks were defined considering the videos and the lumbar motion from the elgons.

The duration of the phases was determined and frame analysis was performed for those. The kinematic data (joint angles) was measured on video frames using the Angles software version 1.2 (University of Alberta). The kinematic information (joint angles from the videos), the height, weight, age and gender of the nurses; the forces used, and the vector direction were used as input to estimate instantaneous compressive and shear forces on the lumbar spine (L5/S1) using the software 3D Static Strength Prediction Program ™ (3D SSPP, v. 4.3 University of Michigan, 2000).

The 3D SSPP is a software package including a static, two or three-dimensional biomechanical model of the spine (Bean *et al.*, 1988; Chaffin and Andersson, 1991). The model uses a double-linear optimization approach to provide estimates of moments, compression and shear forces at L5/S1, and estimates of the muscle strength requirements to maintain the system in static equilibrium. The calculation inputs include kinematic data from thirteen joints (ankles, knees, hips, trunk, shoulders, elbows, and wrists), three anthropometric characteristics (height, weight and gender), and six aspects of the external load (weight, position, vectors magnitudes and directions). Detailed description of the biomechanical model in this software is provided elsewhere (Chaffin and Erig, 1991). The 3D SSPP was used to estimate the ligament strain and the population without sufficient torso strength to perform the tasks and phases by comparing the muscle strength requirements with an extensive database of strength capability data for the US adult population (Chaffin and Andersson, 1991).

The validity of the 3D SSPP software estimates was previously established (Chaffin and Andersson, 1991). The estimated compression forces were evaluated in relation to the National Institute for Occupational Safety and Health (NIOSH) action limit value for compression at L5/S1 intervertebral disk (3400 N) (NIOSH, 1981; Waters *et al.*, 1993). In addition to the instantaneous forces, the cumulative compression and shear forces during the tasks were calculated. The cumulative forces were calculated according to Kumar (1990) by multiplying the load of the different tasks and phases by their duration.

Due to ethical considerations, nurses played the role of the "patient" during the recorded transfers. The "patients" were instructed not to assist during the transfers to best simulate actual patients. The same nurses participating in the study exchanged the "patient" role. The forces were determined by job simulated tests presented elsewhere (Vieira and Kumar, 2006a). It has been previously found that workers can reliably simulate the forces exerted during the job in common tasks (ICC 0.75-0.95) (Kumar, 1993; van der Beek *et al.*, 1999; Wiktorin *et al.*, 1996).

11.2.5. Tasks

The tasks performed were patient transfers from stretcher to bed (S-B), bed to stretcher (B-S), bed to chair (B-C), chair to bed (C-B), chair to wheelchair (C-W), wheelchair to chair (W-C), turning patients toward (T-T) and away (T-A) from the nurse side of the bed, and repositioning patients up in bed (U-B) towards the headboard. Figure 11-2 illustrates the tasks analyzed.

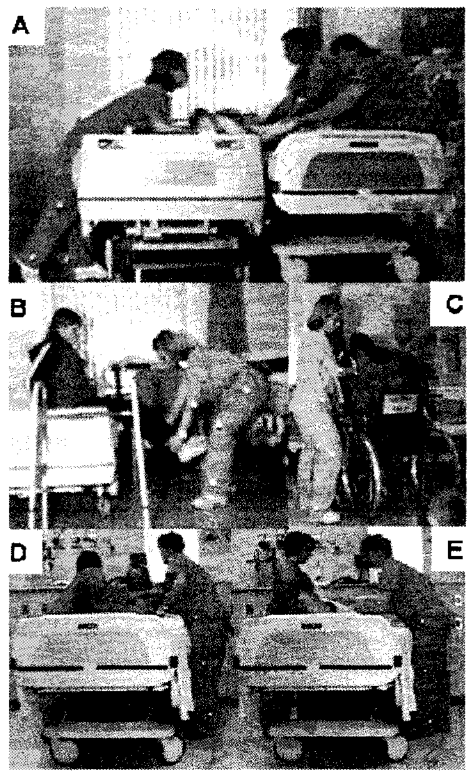

Figure 11-2. Nursing tasks analyzed: A = patient transfer from stretcher to bed; B = patient transfers from bed to chair; C = patient transfers from wheelchair to chair; D = repositioning patient up in bed; E = patient turning.

S-B and B-S transfers involved two nurses one on each side of the bed. The mean (SD) reported number of this type of transfer per shift was 8 (5). The only assistive devices used during the transfers were sliding boards and sheets. The S-B transfer was divided into three phases: preparation, positioning, and pulling phases. During the preparation phase, the nurse was standing up talking with the other nurses about the procedures they would follow. During the positioning phase, the nurses were bent forward supporting the body with the upper limbs against the bed; 40% of the body weight of the nurses was entered as the vertical bed reaction force (from 259 to 321 N). During the pulling phase, the nurses were bent forward reaching the patient across the bed on the stretcher and pulling her towards the bed. The average between the job simulated pulling and lifting force (186 N). The B-S transfer was also divided into three phases: turning the patient, waiting, and pushing. During the turning phase, the nurses were bent forward turning the patient towards them while the co-workers were positioning the sliding board under the patient. The average job simulated pulling force was entered as the external force (149 N). During the waiting phase, the nurses were standing up waiting for the other nurses to position themselves for the transfer. During the pushing phase the nurses started standing up and pushed the patient from the bed to the stretcher. The external force entered was the average between the job simulated pushing and lifting forces (194 N).

T-T, T-A, and U-B were manual tasks (no assistive devices other than the bed sheet) frequently performed in the ICU to avoid pressure ulcer development. The mean (SD) of this type of tasks per shift was 8 (3). The T-T task was divided into three phases: reaching, pulling towards, and holding. During the reaching phase, the nurses were bent forward across the patient reaching for the sheet on the other side of the bed. During the pulling towards phase, the nurses were bent forward turning the patient towards them. During the holding phase, the nurses were standing upright maintaining the patient on his side while the other nurse was placing pillows to support the patient in that position. The job simulated pulling force was entered as the external force (149 N) during the pulling and holding phases. The T-A task was also divided into three phases: organizing, pushing away, and sustaining. During the organizing phase the nurses were standing up grasping the sheet on their side of the bed. During the pushing away phase, the nurses were bending forward rolling the patient away from her. During the sustaining phase, nurses secured the patient in place while the other nurse placed pillows to support him in that position. The job simulated pushing force (194 N) was entered as the external force for the pushing and sustaining phases. Finally, the U-B task was divided into two phases: bracing and lifting. During the bracing phase, the nurse was standing up getting the sheet on her side of the bed and positioning herself for the lift. No external load was entered. During the lifting phase, the nurse lifted the patient toward the headboard of the bed. The average job simulated lifting force (244 N) was entered as the external force at -45° in relation to the vertical (lifting towards the headboard).

The B-C and C-B transfers involved one nurse using a belt around the patient waist to help griping to support the patient's weight. The mean (SD) of this type of transfer per shift was 11 (6). The C-W and W-C transfers from chair to wheelchair and back were performed the same way as the B-C. The mean (SD) of this type of transfer per shift was 4 (2). The B-C, C-B, C-W, and W-C transfers were not analyzed with the biomechanical model because these tasks were asymmetrical and involved movements in multiple planes. In order to analyze these tasks biomechanical, multiple synchronized cameras would be required. The worksite setup (actual patient rooms) did not allow for such complex arrangement.

11.2.6. Data Analysis

The lumbar motion during the tasks was normalized by the lumbar ROM in each direction (flexion, extension, right and left lateral flexion and rotation). During the data analysis, the videos were inspected in parallel to the elgons recording. Each task was watched, then the marked elgons recording was selected and the peak and average lumbar motion were entered into a spreadsheet. The estimated instantaneous and cumulative L5/S1 compression and shear forces, ligament strain, and the percentage of the population without sufficient torso strength were also entered in a spreadsheet for statistical analysis.

Statistical analysis was performed using the SPSS statistical package (SPSS Inc., Chicago, IL). Descriptive analysis was performed, including measures of central tendency, measures of variability, percentages, and ratios. The normality of the distributions was tested using the Kolmogorov-Smirnov test, and the homogeneity of the variances was checked using Levene statistic. The biomechanical demands (ROM vs. job motion; spinal load on the different tasks and phases, and percent of the population not capable) were compared using analysis of variance (one-way ANOVA) with Fisher's least significant difference (LSD) post hoc test. The significance level was set to 0.05.

11.3. Results

11.3.1. Lumbar range of motion (ROM)

The lumbar ROM of the nurses is presented in Table 11-1. There were no significant differences between contra-lateral movements (left and right lateral flexion and rotation). The highest variations on the ROM were found for lumbar extension and rotation (coefficient of variation ≥60%). For flexion and lateral flexion the variation was lower than 25%.

Table 11-1. Lumbar range of motion of the nurses.

Movement (degrees)	Mean ROM	Standard Deviation	Standard Error	95% CI Lower	95% CI Upper
Flexion	53	8	1	51	55
Extension	15	9	1	13	18
R Flexion	25	6	1	23	26
L Flexion	23	5	1	22	25
R Rotation	17	13	2	14	21
L Rotation	18	11	1	15	21

11.3.2. Peak lumbar motion and average lumbar posture during the tasks

Peak flexion during the transfers was higher than during turning and repositioning patients in bed (mean difference > 6°, $p < 0.05$). There were no significant differences on the peak lumbar extension between tasks. The average lumbar flexion during the stretcher to bed transfer was higher than during all other tasks (mean difference > 6°, $p < 0.04$). Figure 11-3 presents the normalized peak lumbar motion during the tasks. The percentage of flexion ROM was higher than the percentage of ROM required for all other movements (mean difference > 15%, $p < 0.001$). The normalized peak flexion was no different than the flexion ROM for the stretcher to bed, bed to chair, and chair to bed transfers.

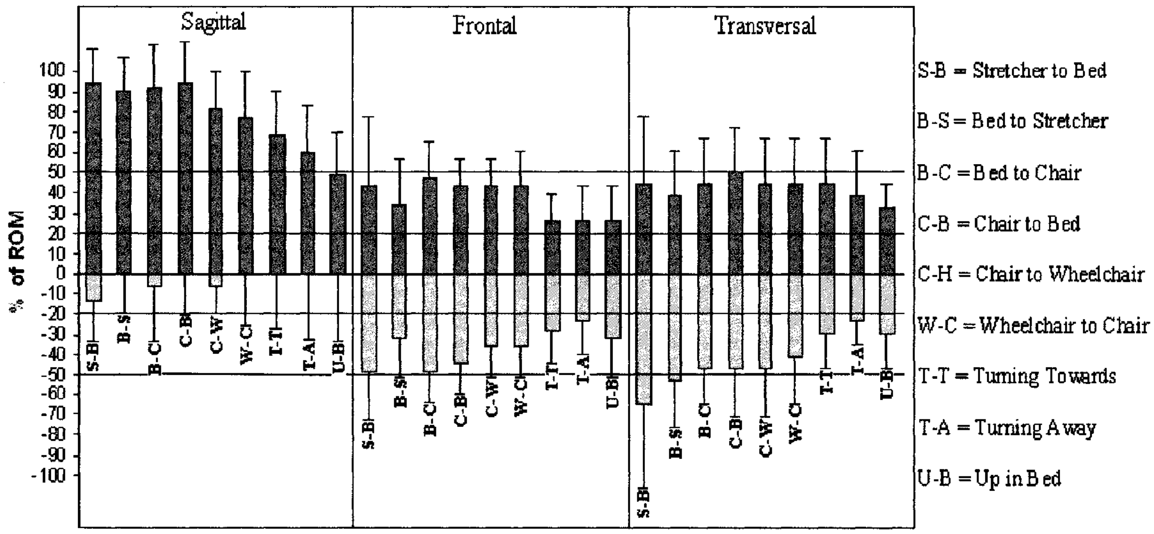

Figure 11-3. Peak lumbar motion during the tasks as a percentage of the maximum range of motion (% of ROM) in the sagittal, frontal and transversal planes (positive deflection = flexion, left lateral flexion, and left rotation; negative deflection = extension, right lateral flexion, and right rotation).

The peak flexion was not higher than 50% of the ROM only during the U-B task (mean difference > 11%, $p < 0.02$). For the other movements, only the peak right rotation during the S-B and B-S transfers were higher than 50% of the ROM. The normalized peak extension during the S-B and B-C was higher than during the B-S and C-B transfers, respectively (mean difference > 7%, $p < 0.032$). The peak right flexion during the S-B was higher than during the B-S transfer (mean difference = 17%, $p = 0.002$). The peak left flexion and right rotation during the transfers was higher than during the turning and repositioning in bed tasks (mean difference > 13%, $p < 0.05$). For the normalized peak left rotation, the difference between the tasks was only significant for repositioning the patient up in bed (lower) in relation to the C-B transfer (mean difference = 15%, $p = 0.011$). In general, the normalized lumbar posture was flexed in the sagittal plane (mean posture > 20% of flexion ROM), but close to neutral posture in the frontal and transversal planes. Lumbar extension and movements in the frontal and transversal planes during the tasks were small, but all tasks were performed with average lumbar flexion higher than 20% of the ROM, and the peak flexion movement was higher than 50% of the ROM in most cases. Figure 11-4 presents the normalized average lumbar posture during the tasks.

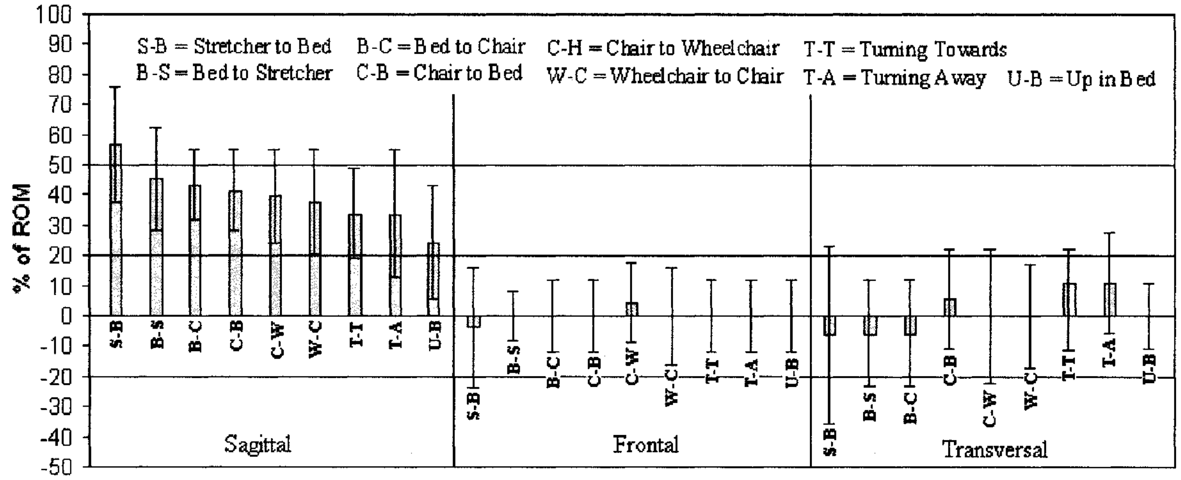

Figure 11-4. Average lumbar posture during the tasks as a percentage of the maximum range of motion (% of ROM) in the sagittal, frontal and transversal planes (positive deflection = flexion, left lateral flexion, and left rotation; negative deflection = extension, right lateral flexion, and right rotation).

The normalized average lumbar flexion during the stretcher to bed transfer was higher than 50% of the ROM and than the average posture during all other tasks (mean difference > 12%, $p < 0.01$). The normalized average lumbar flexion was closest to neutral when repositioning the patients up in bed ($p < 0.03$). There were no differences between the normalized average lumbar postures in the frontal plane during the tasks, and the average lumbar posture in the transversal plane (left flexion) was higher when turning patients (away and towards the nurse) than during the other tasks ($p < 0.05$).

11.3.3. Compression and shear forces at L5/S1 during the tasks and phases

The instantaneous compression forces at L5/S1 were higher when repositioning patients up in bed (mean difference > 749 N, $p < 0.001$) and during the B-S transfer (mean difference > 450 N, $p < 0.017$) than during the other tasks. The instantaneous shear forces at L5/S1 were higher when repositioning patients up in bed than during the other tasks (mean difference > 27 N, $p < 0.032$, not significant in relation to turning towards). The S-B (mean difference > 11 s, $p < 0.001$) and B-S (mean difference > 7 s, $p < 0.001$) transfers took longer to complete resulting in higher cumulative compression (mean difference > 4303 N, $p < 0.001$) and shear forces (mean difference > 351 N, $p < 0.005$).

The positioning phase of the stretcher to bed transfer was the longest of all phases (mean = 21 s, SD = 5 s, mean difference > 11 s, $p < 0.001$) resulting in higher cumulative compression forces at L5/S1 (mean = 24,454 N, SD = 12,084 N, mean difference > 8400 N, $p < 0.001$). The turning (mean = 2374 N, SD = 560 N, mean difference > 395 N, $p < 0.023$) and waiting (mean = 2361 N, SD = 1042 N, mean difference > 383 N, $p < 0.028$) phases of the bed to stretcher transfer resulted in higher cumulative shear forces. Table 11-2 presents the duration and lumbar loads during the tasks.

Table 11-2. Duration and lumbar loads during the nursing tasks. B-S: transferring from bed to stretcher, S-B: transferring from stretcher to bed, T-A: turning away, T-T: turning towards, U-B: repositioning up in bed.

Variable	Task	Mean	Standard Deviation	Standard Error	95% CI Lower	95% CI Upper
Duration (s)	B-S	20	6	1	17	23
	S-B	32	8	2	28	35
	T-A	11	3	1	10	13
	T-T	12	3	1	11	14
	U-B	7	3	1	6	9
Instantaneous Compression (N)	B-S	2457	1691	230	1995	2918
	S-B	1986	1263	172	1642	2331
	T-A	2006	586	62	1884	2129
	T-T	1641	531	56	1530	1752
	U-B	2756	1363	176	2404	3108
Instantaneous Shear (N)	B-S	314	55	7	299	329
	S-B	271	171	23	224	317
	T-A	186	45	5	176	195
	T-T	322	44	5	313	332
	U-B	350	111	14	322	379
Cumulative Compression (N)	B-S	12577	5857	797	10979	14176
	S-B	14493	10381	1413	11660	17327
	T-A	7214	4506	475	6271	8158
	T-T	6584	4614	486	5618	7551
	U-B	8273	3957	511	7251	9296
Cumulative Shear (N)	B-S	1974	909	124	1726	2223
	S-B	1667	767	104	1458	1876
	T-A	699	500	53	594	804
	T-T	1315	738	78	1161	1470
	U-B	1150	591	76	998	1303

Figure 11-5 presents the instantaneous compression and shear forces during the phases of the nursing tasks. Higher instantaneous compression forces were estimated for the pushing phase of the bed to stretcher transfer (mean difference > 2340 N, $p < 0.001$), for the lifting phase of repositioning patients up in bed (mean difference > 1634 N, $p < 0.001$), and for the pulling phase of the stretcher to bed transfer (mean difference > 1208 N, $p < 0.001$) than for the other phases. The same pulling (mean difference > 134 N, $p < 0.001$) and lifting (mean difference > 100 N, $p < 0.001$) phases, respectively, resulted in higher shear forces on L5/S1.

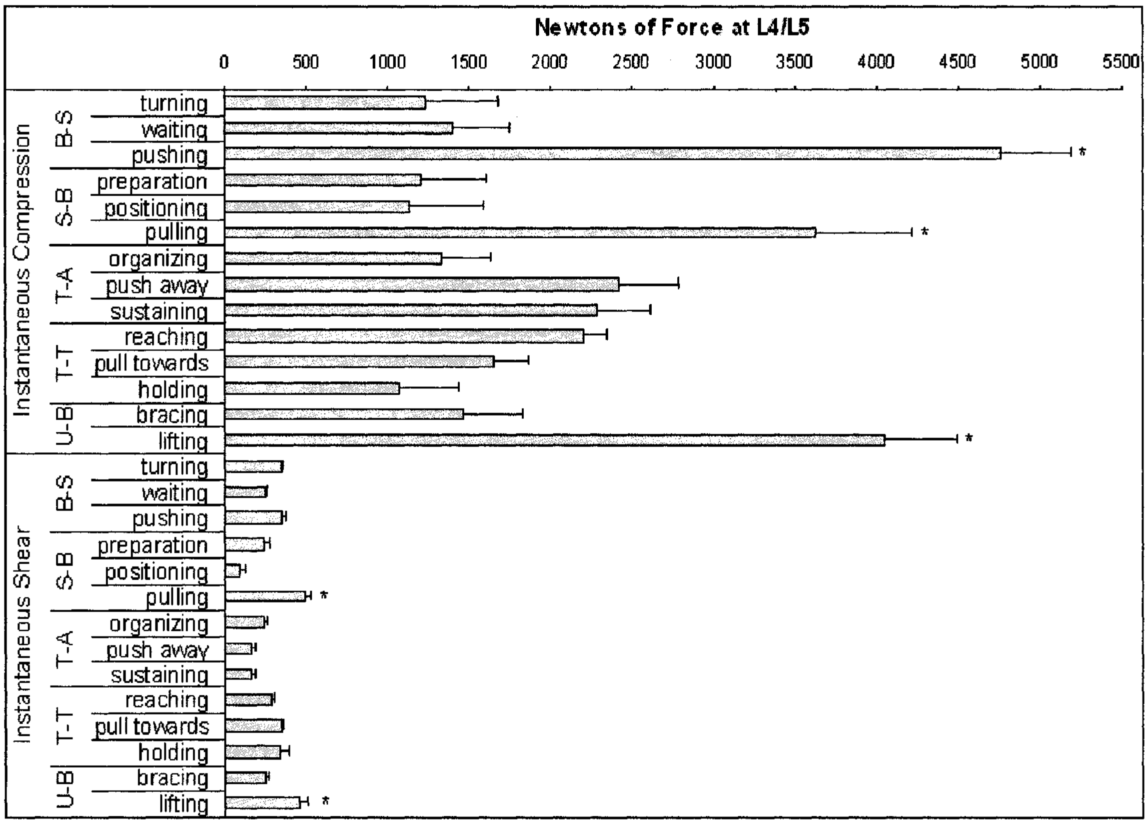

Figure 11-5. Instantaneous compression and shear forces during the phases of the nursing tasks – B-S: transferring from bed to stretcher, S-B: transferring from stretcher to bed, T-A: turning away, T-T: turning towards, U-B: repositioning up in bed.

11.3.4. Percentage of predicted ligament strain and percentage of the population without sufficient torso strength to perform the tasks and their phases

Predicted ligament strain was highest during the stretcher to bed transfer (mean = 14%, 95%CI 5-19, mean difference > 1.3%, $p < 0.041$). It was also estimated that higher percentages of the population would not have sufficient torso strength to perform this task (means = 17%, 95%CI 13-22, mean difference > 9%, $p < 0.001$), to reposition the patients up in bed (mean = 15%, 95%CI 12-18, mean difference > 7%, $p < 0.001$) and to transfer patients from bed to stretcher (mean = 14%, 95%CI 10-19, mean difference > 6%, $p < 0.001$) than to perform the turning tasks. Figure 11-6 presents the percent of ligament strain and percent of the population without sufficient torso strength to perform the different phases of the nursing tasks.

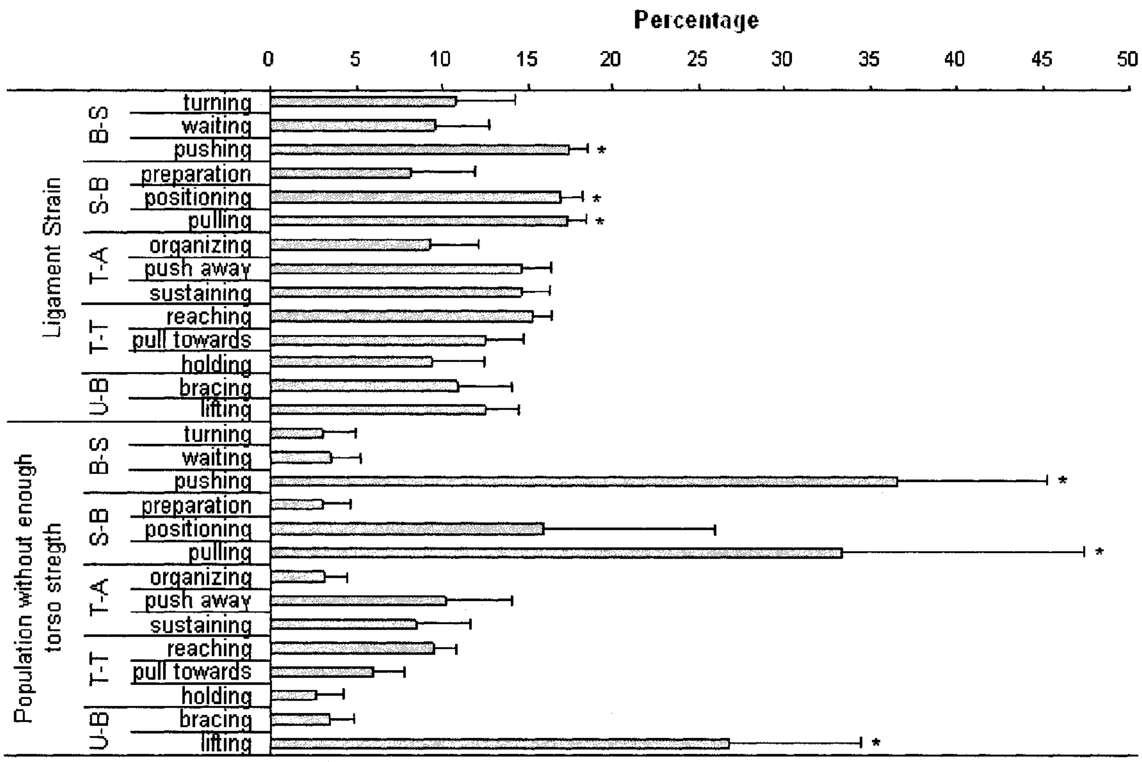

Figure 11-6. Percent of ligament strain and percent of the population without sufficient torso strength to perform the phases of the nursing tasks. – B-S: transferring from bed to stretcher, S-B: transferring from stretcher to bed, T-A: turning away, T-T: turning towards, U-B: repositioning up in bed.

The pushing phase (mean difference > 2.0%, $p < 0.004$) of the bed to stretcher transfer and the pulling (mean difference > 2.0%, $p < 0.004$) and positioning (mean difference > 1.6%, $p < 0.022$) phases of the stretcher to bed transfer resulted in higher percentages of ligament strain than the other phases. The same pushing (mean difference > 20%, $p < 0.001$) and pulling (mean difference > 17%, $p < 0.001$) phases, and the lifting phase (mean difference > 10%, $p < 0.001$) of repositioning patients up in bed were the phases that the highest percentage of the population would not have sufficient torso strength to perform.

11.4. Discussion of Chapter 11

The data presented contributes to advance the knowledge in the functional capacity of nurses; ROM data of nurses with work-related low back disorders (WLBD) may be compared with the healthy nurses ROM data presented and it may be useful to assess treatment efficacy. In general, peak motion during the tasks performed by the orthopedic nurses (ON transfers) was higher than during the tasks by the ICU nurses (IN turning and repositioning). This may partially explain the higher rates of WLBD among ON than IN. The average flexion and percentage of ligament strain during the stretcher to bed (S-B) transfer was higher than during all other tasks showing that this is a high risk transfer for the nurses. In addition, the full flexion ROM was used during the S-B, bed to chair (B-C), and chair to bed (C-B) transfers exposing the nurses to increased risk of overexertion injuries during these frequent ON tasks. The motion during the job should be within 20% of the ROM and not exceed 50% of the ROM (Kumar, 1994; Vieira and Kumar, 2004a and 2004b). The National Institute for Occupational Health and Safety (NIOSH) suggested even more stringent guidelines with an action limit for joint motion of 10% of the ROM and a maximal permissible limit of 30% of the ROM (NIOSH, 1981). Thus, nursing tasks (specially the S-B, B-C, and C-B transfers) expose the workers to high risk of WLBD due to the large motions associated with force exertion. The fact that maximum flexion was used during 3 out of the 9 tasks analyzed showed that the concern about overstretching the elgons during flexion ROM testing was overzealous because the elgons were tested periodically during the data collection period with no decrease in accuracy.

The biomechanical model used in this study assumed that the system was in static equilibrium, for these reason the lumbar loads during the tasks may be even higher and the percentage of the population capable of performing the transfer may be lower than presented because the tasks had dynamic components. Despite, the compression forces were higher than the NIOSH action limit (3400 N) during the pushing phase of the bed to stretcher (B-S) transfer, during the lifting phase of repositioning patients up in bed (U-B) and during the pulling phase of the S-B transfer.

In a previous study, the peak compression load of workers who reported WLBD (3423 N) was higher than for the group who did not report (2733 N, $p < 0.001$) (Norman *et al.*, 1998). The shear force in the WLBD group was 462 (178) N and resulted in a probability of being in the WLBD group of between 50% and 60%. The instantaneous shear forces during the pulling phase of the S-B transfer (mean = 487 N, SD = 40) and during the lifting phase of the U-B task (mean = 454 N, SD = 53) were similar to the values reported for workers with WLBD. The combination of high bending moments, shear and compression forces, such as the ones we found during the manual transfers, increases the risk of injury to the lumbar intervertebral disks (Shirazi-Adl, 1989).

The action limit proposed by NIOSH for instantaneous compression forces are useful to evaluate risk of WLBD; however even forces below the action limit may expose the workforce to high cumulative loads and the action limit "may not protect the entire workforce" (Vieira and Kumar, 2006b, Waters *et al.*, 1993). The time to complete each task was directly related to the resulting cumulative forces. No cut-points for compression and shear taking into account for the cumulative effect of the exertions on the viscoelastic tissues of the body were found. Peak and cumulative loads are both important, independent risk factors for WLBD (Kumar, 1990; Norman *et al.*, 1998). Cumulative compression and shear forces were shown to be higher in institutional aids with low back pain than in those without pain (Kumar, 1990).

One of the limitations of this study is that the cumulative load estimation method used may have introduced some errors because not all frames were modeled to estimate the loads. In addition, we were not able to analyze the bed to chair to bed and chair to wheelchair to chair transfers with the biomechanical model because the nurses moved out of the video-recording plane. Thus, we could not estimate the compression and shear forces, nor the ligament strain and population without sufficient torso strength to perform these tasks. Future studies could perform laboratory simulations to evaluate these tasks in an environment that would allow for the use of multiple cameras.

Transferring patients from B-S [pushing phase = 63%] and from S-B [pulling phase = 67%], and the U-B [lifting phase = 73%] were tasks acceptable by less than 75% of the population. Workers who performed activities accepted by less than 75% of all workers presented three times more WLBD (Snook, 1978). Training in patient handling and transfers alone is not enough to reduce WLBD incidence (e.g. Hignett, 1996). The use of mechanical assistive devices reduced the compression forces during bed to bed transfers from 2955 to 1189 N (Silvia *et al.*, 2002). Despite reducing the peak load, the use of lifting devices may increase the cumulative load because of the longer time to complete patient transfers (Daynard *et al.*, 2001). Nursing tasks should be analyzed individually because the demands vary even when alternating the direction of the task (e.g. B-S vs. S-B).

Training and education combined with ergonomic interventions, using mechanical lifts, and regular exercise were found to be effective in reducing WLBD rates in nurses (Bos *et al.*, 2006; Garg and Owen, 1994; Hignett, 2001; Li *et al.*, 2004; Skargren and Oberg, 1996). Introducing additional mechanical lifts, minimizing the time to complete the transfers, reducing the amount of trunk flexion and forces required during the transfers will greatly reduce the lumbar loads and risk of WLBD in the nursing tasks analyzed. Training programs could be implemented to emphasize the importance of minimizing the time unnecessarily spent in trunk flexion. When turning the patient toward, if possible, the nurse should flex the patient contra-lateral hip and knee, position her hands on the knee and use it as a hinge to reduce the reach distance and force required. When repositioning the patient up in bed, the nurse should first lower the headboard to reduce the amount of lifting force required. Hopefully the information provided in this paper will advance the knowledge in nursing and the recommendations will contribute to reduce the risk of WLBD.

11.5. Conclusions of Chapter 11

Nursing tasks impose significant biomechanical demands on the lumbar spine contributing to the high incidence of low back disorders observed in this group. The demands of nursing tasks vary depending on the tasks in different hospital departments. Reduction of biomechanical demands is needed for low back disorders control. High flexions and forces are critical aspects of the transfers requiring most of the nurses' capabilities. Potentially beneficial job/task modifications and training programs were determined based on the results of the epidemiological, questionnaire, and physical demands studies, and on the literature. In general, the modifications suggested include design modifications (room size and setup), acquisition of new equipment (mechanical lifts), working technique modifications (to maximize biomechanical advantageous use of the body), and worker education (no smoking and regular exercising campaigns). Thus, the recommendations involve a combination of re-engineering strategies, training and worker education programs. Fitness to work, job modifications and training programs can be designed and assessed based on the results. Further studies could evaluate if interventions designed based on the information gathered using this methodology can successfully reduce the incidence of low back disorders in nurses.

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Chapter 12 – Motion and Biomechanical Demands of the Punching Job of Steel Workers

Following a study that found high annual rates of low back disorders among computer numeric control (CNC) steel workers (5.4%) and found that the punching job was considered the most physically demanding and risky CNC job in the steel company evaluated (Chapter 7), this chapter presents the study that quantified the biomechanical demands on the lumbar spine during the punching job of CNC steel workers. A version of this chapter has been submitted for publication in the Journal of Safety Research (Vieira and Kumar, 2006a).

12.1. Introduction

Work-related low back disorders (WLBD) are the most frequent and most costly musculoskeletal disorder (Woolf, 2003). In 2000, based on the National Health Interview Survey, from 11 to 13 million people developed low back pain, and about $100 billion were spent on this problem in the US. Considering the people who had worked in the previous year, 22.4 million had back pain (National Academy of Sciences, 2001). Between 1991 and 2001, more than 40% of all injures and illnesses resulting in days away from work in the US were sprains and strains (43.6% in 2001) [Bureau of Labor Statistics (BLS), 2001]. The back accounts for approximately 40% of all lost time claims in the US (Guo *et al.*, 1999). In 2001, WLBD represented 24.2% of all lost time claims, 65% of all WLBD were caused by overexertion, and 83% of the WLBD cases (N = 183,424) were classified as sprains and strains (N = 152,505) (BLS, 2001). The male workers with the highest rates of WLBD (40% of all cases) were industry laborers, operators, and fabricators (BLS, 2001). Similarly, between 2001 and 2005, WLBD accounted for 20% to 30% of all work-related compensation claims in Canada [Alberta Human Resources and Employment (AHRE), 2005]. In 2005, overexertion was the most common cause for lost-time claims in all industries in Alberta (AHRE, 2005). Sprains, strains and tears were the leading nature of injury; 75% of the WLBD were classified as sprains, strains, and tears, and approximately 70% of these resulted from overexertion in lifting, pulling, pushing, carrying, twisting, climbing, tripping, or reaching (AHRE, 2005). The back was the body part most frequently injured accounting for over one-third of all claims, and about 27% of all lost time claims accepted by the Workers' Compensation Board of Alberta were WLBD (AHRE, 2005).

WLBD precipitates through the steps of activity, biomechanical stress, temporal loading, tissue strain, and injury (Kumar, 2001; Marras, 2000; McGill, 1997). The relationship between work demands and WLBD has been established by many studies for many years (Bernard, 1997; Frymoyer *et al.*, 1980). There is evidence that awkward postures, heavy work, lifting, forceful movements and whole body vibration have causal association with WLBD (Bernard, 1997). Cumulative load and overexertion can result in injury by disruption of tissue causing WLBD (Kumar, 1990 and 1994; McGill, 1997). Overexertion is "a physical activity in which the level of effort exceeds normal physiological and mechanical (physical) tolerance limits" (Kumar, 1994). Trunk flexion (relative risk – RR of 1.72 for flexion ≥60° for more than 5% of working time/day), trunk rotation (RR of 1.57 for rotation ≥30° for more than 10% of working time/day), and weight lifting (RR of 1.79 for weight ≥25 kg for more than 15 times/day) were found to be risk factors for WLBD (Hoogendoorn *et al.*, 2000). In addition, spinal compression has long been studied as a risk factor for WLBD, and forms the basis of the NIOSH lifting recommendations (Waters *et al.*, 1993). Compression and shear forces are both affected by posture, contraction, and external loads. NIOSH suggested a maximum permissible limit (MPL) for compression at L5/S1 intervertebral disk of 6700 N, and an action limit (AL) value of 3400 N [National Institute for Occupational Safety and Health (NIOSH), 1981]. It was suggested that tasks at the AL level can be handled by 99% males and 75% females safely, but increases the risk of WLBD moderately. While, tasks at the MPL level are hazardous for 75% males and 99% females, and the risk of WLBD is eight times higher than at the AL level (Waters *et al.*, 1993). The MPL is based on in-vitro studies that found micro-fractures on the vertebral cartilage endplates of subjects under 40 years old when applying 6700 N of axial load, and the AL is based on the fact that micro-fractures on the vertebral cartilage endplates of subjects 60 years or older started to happen when applying 3400 N of axial load (Evans and Lissner, 1959; Sonoda, 1962).

The physical demands of the steel workers' job are high and they have a high incidence of WLBD (Dueker *et al.*, 1994; Hildebrandt *et al.*, 1996). About 70% of the workers of the maintenance division of a steel mill had WLBD (Udo and Yoshinaga, 2001). In a study of WLBD in two steel companies (n = 618 workers) the a lifetime incidence rate of lumbar symptoms/100 workers was 66%, and the incidence rate of lumbar symptoms during the previous year and during the previous week were 53% and 25%, respectively (Masset and Malchaire, 1994). Thus, it is important to evaluate physical demands and identify risks to WLBD in order to make evidence based recommendations for job modifications and training programs to reduce the risk of WLBD in steel workers.

12.1.1. Background and objectives

An epidemiological study and a questionnaire survey including 108 steel workers were performed (Vieira and Kumar, 2006d). In the steel company studied, the annual rate of low back disorders was highest among workers performing the punching job (5.4%), and it was higher then the rates reported previously for other professionals in heavy jobs (nurses = 3.3%, lumbermen = 3.3%, and construction workers = 2.8%) (Klein *et al.*, 1984). The working-life rate of reported WLBD and the point prevalence of low back pain were 36% and 16%, respectively. The low back discomfort scores were higher than for all other body parts. The mean (SD) weight manually handled was 35 (11) kg. Perceived exertion was strong 5 (1). Repetitions and duration contributed more to the total effort than postures, movements, and force ($p < 0.044$, 95% CI 0.1 to 8.0). The punching job was considered the most physically demanding and risky job by the steel workers (Vieira and Kumar, 2006d). Following these studies, the forces and electromyographic activity (EMG) of the rectus abdominis and erector spinae during maximum voluntary exertion (MVE), job simulated, and preferred levels for lifting (squat and stoop), pushing, and pulling were measured (Vieira and Kumar, 2006c). The job simulated force [72 (22) % of MVE] was higher than the preferred level of exertion [55 (17) % of MVE] ($p < 0.01$), and it was close to the maximum during pushing (0.8, 95%CI 0.8-0.9). The EMG of the erector spinae during the job simulated pushing was 1.5 times (95%CI 1.3-1.7) higher than at the preferred level (Vieira and Kumar, 2006c).

The objective of this study was to test a methodology and quantify the physical demands of the punching job of steel workers including lumbar motion, compression and shear forces, percentage of ligament strain and percentage of the population with sufficient torso strength to perform the job. This article presents the assessment of the physical demands of the punching job of steel workers, and evidence based suggestion of recommendations for job modifications and training programs.

12.2. Methods

This study was approved by the University Research Ethics Committee for Human Studies. Steel workers from the computer numeric control (CNC) sector of a steel blade manufacturing company participated in the study.

12.2.1. Subjects

Twenty-three CNC steel workers with no low back disorders resulting in time off work during the previous twelve month participated in the study. The volunteer workers received further explanations about the objectives and procedures of the study, including a statement of their right not to participate and to withdraw at any time with no consequence to them, and signed a consent form. The mean (standard deviation) age of the steel workers analyzed was 27 (5) years; the height was 178 (1) cm; the body mass was 83 (13) kg, and the body mass index was 26 (4) kg/m^2.

12.2.2. Punching job

The participating steel company processes blades for tractors, snowplows and other machines. The punching job required to program, set clamps, load, tighten up and untighten blades to machines, unload machines manually or using cranes; the machines punched holes on the blades to fit screws and bolts. The blades were pushed or pulled on rollers while on machine tables. The CNC steel workers manually aligned steel blades on the conveyor belt so that the punch machine press-stud lined up with the pre-drilled hole, then the machine was activated and the hole was punched in the blade. Several holes were punched in each blade.

The average [mean (SD)] weight and dimensions of the steel blades were 35 (11) kg, 99 (31) cm long, 25 (5) cm wide, and 3 (0.6) cm thick. Data was recorded while holes were punched in three blades. The work shift was 12 h/day, 42 h/week; there was monthly rotation between day and night shifts.

12.2.3. Lumbar range of motion: procedures and data collection

In order to normalize the data and allow the calculation of the thresholds for analysis, the maximum lumbar range of motion (ROM) of the subjects was assessed. Three maximum flexion, extension, lateral flexion (right and left), and rotation (right and left) trials were performed to calculate the mean lumbar ROM in each direction. The sequence of the trials and directions was randomized. The subjects were instructed to "bend forward/backward as far as you can", "bend to the right/left side as far as you can" and to "rotate to the right/left side as far as you can".

The lumbar ROM on the frontal and transversal planes was measured using electrogoniometers (one goniometer XB 180 and one torsiometer Q1 50 – Biometrics, Gwent, UK). The goniometer permits measurement of angular movement of both sagittal (flexion/extension) and frontal (left/right side flexion) planes, and the torsiometer permits measurement of angular movements in the transversal plane (left and right rotation), independently of the linear movement between the two endblocks (Biometrics, 1999). The devices will be referred as elgons. The lumbar spine was palpated and the spinal processes of T12 and L5 were marked using a dermatologic pencil. The endblocks of the elgons were placed in attachment ducts and were anchored to the skin over the landmarks using double-adhesive tape (Vieira and Coury, 2002). Single-adhesive tape was put on top, and the connection cables were attached to the elgons and respective channels on the data link. The lumbar ROM on the frontal plane [lateral flexion: left (+) and right (-)] was recorded on Channel 1 and the lumbar ROM on the transversal plane [rotation: left (+) and right (-) was recorded on Channel 3. The data was transferred to a laptop computer using the Biometrics Data Link (Biometrics, Gwent, UK).

According to the manufacturer's manual, the operating temperature of the elgons is from 0° to 40°C; the operating humidity range is from 30% to 75%; the repeatability error of is $\leq 1°$, and the maximum measurement error is ± 3° and ± 5° when measuring single and multiple planes of movement, respectively, within 60° of motion (Biometrics, 1999). The small repeatability and measurement errors of the back elgons were confirmed by independent studies (Shiratsu and Coury, 2003; Vieira and Coury, 2002).

The elgons were not used for sagittal ROM measures due to the potentially hazardous stretching of the devices. The extreme flexion ranges could overstretch the wire that keeps the endblocks of the elgons together and contains the strain gauges that measure the angular displacement (Biometrics, 1999). The lumbar ROM in the sagittal plane was measured using perpendicular markers photogrammetry (Vieira and Coury, 2004). This method has been used to measure the lumbar posture and sagittal motion at least since 1974 (Kumar, 1974). The concurrent validity of this method to measure the orientation of lumbar segments was found to be good when comparing to X-ray measurements (r^2 from 0.91 to 0.98) (Chen and Lee, 1997). The parallel reliability of sagittal plane measures of lumbar motion using perpendicular markers photogrammetry was also found to be high when compared with elgons' measures [$r^2 = 0.994$; measurement error: 0.6° (SD = 0.7)] (Vieira and Coury, 2004).

The perpendicular markers were placed on the skin over the landmarks using double-adhesive tape and single- adhesive tape was put on top. The subjects were video-recorded from the sagittal plane (right side) to measure the flexion and extension ROM. Thirteen frames were extracted from video using windows movie maker to represent the neutral posture (seven frames), fully flexed posture (three frames), and fully extended postures (three frames). An audio video interleave (AVI) movie was created with the thirteen jpeg pictures using jpegAvi software. The new avi video was resaved using the VirtualDub software to allow for analysis using the PostureAna software (University of Alberta). The origin and tips of the perpendicular markers were marked clockwise on the thirteen frames and the angles were calculated.

The difference between the neutral posture and the fully flexed posture defined the flexion ROM and the difference between the neutral and fully extended postures defined the extension ROM. Figure 12-1 illustrates the measure of the lumbar ROM in the sagittal plane using perpendicular markers photogrammetry.

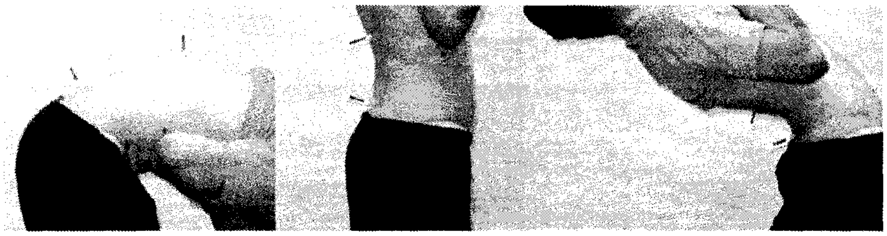

Figure 12-1. Lumbar range of motion measurement in the sagittal plane using perpendicular markers photogrammetry.

12.2.4. Lumbar motion during the punching job: procedures and data collection

The lumbar motion in the sagittal, frontal, and transversal planes during the punching job was recorded using elgons. The elgons were fixed leveled with T12 and L5. Once the elgons were attached, the worker performed the job. The data was transferred to a computer using the Biometrics Data Link and the peak and average lumbar motion during the punching job was determined using the elgon software (Biometrics, Gwent, UK). The recording included the worker punching multiple holes in three blades. The lumbar motion on the frontal plane [left (+) and right (-) lateral flexion] was recorded Channel 1; the lumbar motion on the sagittal plane [flexion (+) and extension (-)] was recorded on Channel 2; the lumbar motion on the transversal plane [left (+) and right (-) rotation] was recorded on Channel 3. In addition, an event marker was recorded on Channel 8. The event marker switch activated a red light LED which appeared in the corner of the foreground of the film and generated a spike on Channel 8 of the data logger. Thus, the video-recordings and elgons data were synchronized by means of a LED and an electric pulse (step increase in voltage) recorded along with the elgons data.

12.2.5. Lumbar compression and shear forces, percentage of ligament strain and percentage of the population capable of performing the tasks: procedures and load estimation

The punching job was video-recorded using a Canon mini DV media digital camera, at 30 frames per second, from the sagittal plane (workers' left side), while the 23 CNC steel workers punched holes in three steel blades. The videos were used to measure the postures and to determine beginning, end, and duration of the job tasks. The location of the joints was estimated to link body segments; surface markers were used to facilitate joint location identification. The job tasks were defined considering the videos and the lumbar motion from the elgons. The duration of the tasks was determined and frame analysis was performed for those. The joint angles were measured on video frames using the Angles software version 1.2 (University of Alberta). The kinematic information, the height, weight, age and gender of the workers; the forces used, and the vector direction were used as input to estimate instantaneous compressive and shear forces on the lumbar spine (L5/S1) using the software 3D Static Strength Prediction Program ™ (3D SSPP), version 4.3, 2000 (University of Michigan). The 3D SSPP is a software package including a static, two or three-dimensional biomechanical model of the spine (Anderson *et al.*, 1985; Bean *et al.*, 1988; Chaffin and Andersson, 1991). The model uses a double-linear optimization approach to provide estimates of moments, compression and shear forces at L4/L5 and lumbosacral joints, and estimates of the muscle strength requirements to maintain the system in static equilibrium. The validity of the 3D SSPP software estimates was previously established (Chaffin and Andersson, 1991). Detailed description of the biomechanical model in this software is provided elsewhere (Chaffin and Erig, 1991).

Two frames representing the tasks of positioning the blade and punching the blade were analyzed for each of the three blades punched (six frames per worker, 23 workers, for a total of 138 frames). The forces used were determined during job simulated tests presented elsewhere (Vieira and Kumar, 2006c). It has been previously found that workers can reliably simulate the forces exerted during the job in common tasks (ICC 0.75-0.95) (Kumar, 1993; van der Beek *et al.*, 1999; Wiktorin *et al.*, 1996).

The external force entered for the positioning phase was the mean job simulated pushing forward force (182 N) and for the punching phase was 20% of the body weight (from 137 N to 188 N depending on the worker weight) in the vertical upward direction because the workers were slightly flexed using their upper limbs against the machine to partially support their weight (reaction force). The cumulative forces were calculated according to Kumar (1990) by multiplying the load of the different tasks by their duration (Kumar, 1990). Figure 12-2 shows a steel workers performing the positioning and punching phases of the job; the lower part represents the posture and reaction force vectors in the 3D SSPP.

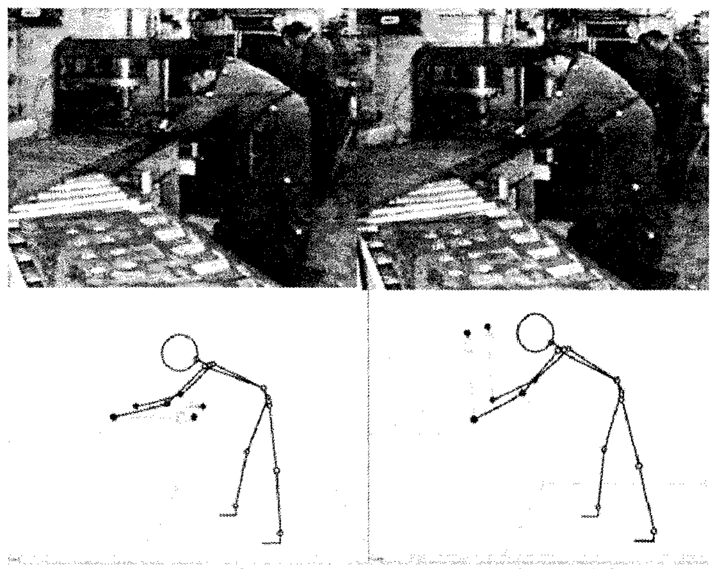

Figure 12-2. Positioning and punching phases of the steel working job.

The percentage of ligament strain and percentage of the population capable of performing the tasks were also calculated using the 3D SSPP. The 3D SSPP estimates the percentage of the population capable of performing the tasks by comparing the muscle strength requirements with an extensive database of strength capability data for the US adult population (Chaffin and Andersson, 1991).

12.2.6. Data analysis

The lumbar motion during the tasks was normalized by the lumbar ROM in each direction (flexion, extension, right and left lateral flexion and rotation). During the data analysis, the videos were inspected in parallel to the elgons recording. Each task was watched, then the marked elgons recording was selected and the peak and average lumbar motion were entered into a spreadsheet. The estimated instantaneous and cumulative L5/S1 compression and shear forces, ligament strain, and the percentage of the population without sufficient torso strength were also entered in a spreadsheet for statistical analysis. Statistical analysis was performed using the SPSS statistical package (SPSS Inc., Chicago, IL). Descriptive analysis was performed, including measures of central tendency, measures of variability, percentages, and ratios. The normality of the distributions was tested using the Kolmogorov-Smirnov test, and the homogeneity of the variances was checked using Levene statistic. The biomechanical demands (ROM vs. job motion; spinal loads, and percent of the population capable) were compared using analysis of variance (one-way ANOVA) with Fisher's least significant difference (LSD) post hoc test. The significance level was set to 0.05.

12.3. Results

There were no significant differences between right and left flexion ROM or between right and left for rotation ROM. The lumbar ROM of the steel workers is presented in Table 12-1. The peak lumbar motion during the punching job is presented in Table 12-2.

Table 12-1. Lumbar range of motion of the steel workers (n = 23).

Movement (degrees)	**Mean ROM**	**Standard Deviation**	**Standard Error**	**95% CI Lower**	**95% CI Upper**
Flexion	49	17	2	45	54
Extension	11	6	1	10	13
R Flexion	26	5	1	24	27
L Flexion	24	4	0	23	25
R Rotation	13	4	1	11	14
L Rotation	12	5	1	10	13

Table 12-2. Peak lumbar motion during the punching job (n = 23 steel workers).

Movement (degrees)	Mean ROM	Standard Deviation	Standard Error	95% CI Lower	95% CI Upper
Flexion	44	6	1	42	46
Extension	0	1	0	0	1
R Flexion	9	5	1	8	11
L Flexion	8	5	1	6	9
R Rotation	5	3	0	4	5
L Rotation	10	3	1	9	12

There were no significant differences between right and left peak lateral flexion, but peak left rotation was higher than the peak right rotation (mean difference = 6°, p = 0.001). The average lumbar posture was in flexion (mean = 33°, SD = 8°), neutral lateral flexion (mean = 1°, SD = 5°) and neutral rotation (mean = -2°, SD = 3°). Figure 12-3 presents the normalized peak and average lumbar motion during the punching job.

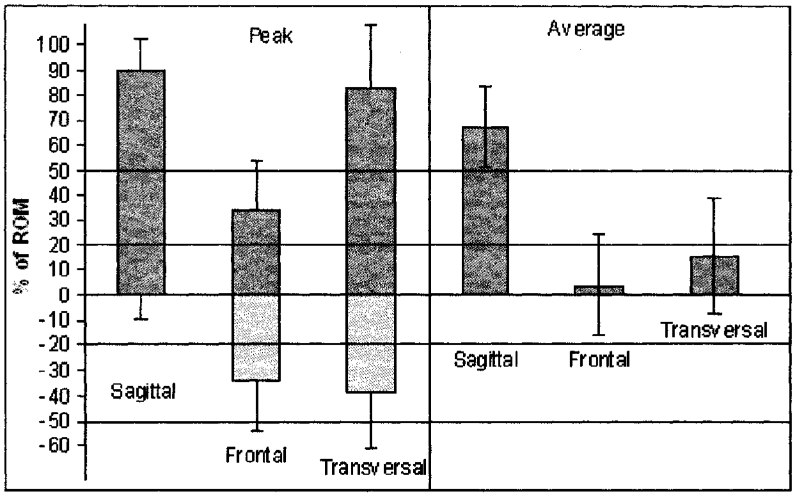

Figure 12-3. Peak lumbar motion and average lumbar posture during the punching job as a percentage of the maximum range of motion (% of ROM) in the sagittal, frontal and transversal planes (positive deflection = flexion, left lateral flexion, and left rotation; negative deflection = extension, right lateral flexion, and right rotation).

Peak flexion and left rotation during the punching job required the highest percentages of the ROM (> 80% of the ROM, $p < 0.001$), and the average lumbar flexion represented more than 60% of the flexion ROM. Table 12-3 presents the lumbar loads and percentage of the population with sufficient torso strength to perform the tasks.

Table 12-3. Lumbar loads during the punching job (n = 23 steel workers).

Variable	Task	Mean	Standard Deviation	CV (%)	Standard Error	95% CI Lower	95% CI Upper
Duration (s)	Positioning	47	25	52	3	40	54
	Punching	39	17	44	2	34	43
Instantaneous Compression (N)	Positioning	2828	318	11	44	2739	2916
	Punching	281	112	40	15	250	312
Instantaneous Shear (N)	Positioning	219	33	15	5	209	228
	Punching	174	11	6	1	171	177
Cumulative Compression (N)	Positioning	134436	69766	52	9675	115013	153859
	Punching	10250	5957	58	826	8592	11909
Cumulative Shear (N)	Positioning	10274	5170	50	717	8835	11714
	Punching	6710	2957	44	410	5887	7533
Ligament Strain (%)	Positioning	16	2	13	0	15	16
	Punching	13	2	16	0	13	14
Population Capable (%)	Positioning	95	1	1	0	95	96
	Punching	99	0	0	0	99	99

Each blade took approximately 1.5 minutes to be punched. The percentage of the population with sufficient torso strength to perform the positioning phase was lower than to perform the punching phase; all other variables presented higher values during the positioning phase. For example, the instantaneous compression force during the positioning phase was 10 times higher than during the punching phase (95% CI 9-11). The difference between the positioning and punching phases was significant for all variables (p = 0.039 for duration of the tasks and $p < 0.001$ for the other variables).

12.4. Discussion of Chapter 12

As expected there were no significant differences in the ROM between contra-lateral movements (right and left lateral flexion or right and left rotation). The ROM data presented contributes to the establishment of databases on the functional capacity of steel workers. Future studies are necessary because our sample was small to allow generalizations for the steel workers' population. Despite, the healthy workers ROM data presented may be useful during disability assessments to compare with the ROM data of steel workers with WLBD, and it may also be useful to assess treatment efficacy.

The left rotation during the punching job was higher than the right rotation; this may be explained by the fact that steel blades come to the punching workstation from the left side requiring higher rotations in that direction to bring the blades toward the punching site. WLBD was found to be associated with maximum flexion angle (odds ratio – OR = 2.2), peak spinal loads (OR = 2.0), average spinal loading (OR = 1.7), percent of time with loads in the hands (OR = 1.5), and percent of time spent working in flexion > 45° (OR = 1.3) (Neumann *et al.*, 2001). An odds ratio of 4.28 for shoulder and back injuries was found for workers using more than 34% of the ROM during work (Punnett *et al.*, 1991). Another study found that, the risk of WLBD increases significantly with prolonged trunk flexion and rotation (OR = 4.9 for mild trunk flexion, and 5.7 for severe trunk flexion, and 5.9 for trunk twist or lateral bend) (Punnett *et al.*, 1991).

The methodology used to measure and evaluate the lumbar motion during the job is unique because it normalizes the motion by the individual range of motion. This approach has been suggested previously, but it is not frequently used in field studies. Raw values are frequently presented as opposed to normalized values. The use of normalized values is important because when the demands exceed the workers capacity injuries may occur (Kumar, 1994). Based on epidemiological studies, it has been suggested that the motion during the job should be within 20% of the maximum range of motion (ROM) and it should not exceed 50% of the ROM (Vieira and Kumar, 2004). Thus, the high peak and average flexion and peak left rotation during the punching job (> 50% of the ROM) representing increased risk of WLBD.

The instantaneous compression forces during the punching job were below NIOSH action limit of 3400 N (NIOSH, 1981).However, the tasks are performed in less than optimal conditions, with increased duration and high frequency. The action limit value "may not protect the entire workforce" (Waters *et al.*, 1993). Similarly to our results (annual rate of 5.4%), an earlier study also found WLBD rates of 5% in 411 workers with the estimated compressive force at L5/S1 higher than 2500 N (Chaffin and Park, 1973). The action limit proposed by NIOSH for compression forces are useful to evaluate risk of WLBD; however they may expose the workforce to high demands (Vieira and Kumar, 2006b).

The static biomechanical model used assumes that the system is in static equilibrium and may have underestimated the loads for the dynamic components of the job (Wiktorin *et al.*, 1996). For these reasons, the lumbar loads during the punching job may be even higher, and the percentage of the population capable of performing the tasks may be lower than presented. In addition, the NIOSH cut-points for compression do not take into account the cumulative effect of the exertions on the viscoelastic tissues of the body (van Dieen *et al.*, 1994). Peak and cumulative loads are both important and independent risk factors for WLBD (Norman *et al.*, 1998). Future studies could be performed to determine cut-points to cumulative forces on the spine. A limitation of this study was that the cumulative load estimation method used may have introduced errors because not all frames were modeled to estimate the loads. Despite, the phase component weighting method used allowed for the evaluation of the load associated with different phases of the task, and allowed for the identification of differences between peak and cumulative loads among the phases, without been prohibitively labor and time intensive. Another potential limitation on the load estimation was that the videos were taken from the sagittal plane, thus asymmetries possible affecting the lumbar load may have been missed. However, we believe that inaccuracies from this source are not significant in our results because both observation of task and the lumbar motion analysis (elgons) results showed that the task was mainly symmetrical and performed in the sagittal plane.

The depth of the punching tables was significant (1 m) requiring increased trunk flexion to reach and position the steel blades. The height of the punching machine was low (1.65 m) and did not provide enough clearance for the head, requiring the workers to stay in trunk flexion for prolonged periods. The duration of the tasks was directly affected by the number of holes to be punched (3 to 6 sec/hole) and on the size of the blades. The time to process each blade was directly related to the resulting cumulative forces. The cumulative forces had higher variation (mean CV = 51%) than the instantaneous forces (mean CV = 18%). Bigger blades required more time to be processed not only because they had more holes to be punched but also because the time to position the blades was longer due to the increased weight. Minimizing the positioning time, reducing the amount of trunk flexion and pushing force required during this task will greatly reduce the lumbar loads and risk of WLBD in the punching job.

The punching table could be elevated to reduce the amount of flexion required and increase the head-room allowing the worker to get closer to the table. Training programs could be implemented to emphasize the importance of minimizing the time unnecessarily spent in trunk flexion such as during the punching phase where some workers return to a more neutral posture but others remain in increased flexion resulting in increased compression (CV = 40%). Finally, the inclusion of rolling spheres that facilitates the positioning of the blades on the punching table would result in decreased use of force during the positioning phase of the punching job.

12.5. Conclusions of Chapter 12

This combined methodology is useful to identify the critical tasks and specific risks for low back disorders in jobs with risk factors. The percent of the population capable of dealing with the instantaneous demands of the punching job was higher than 75%. In addition, the percent of ligament strain was lower than 20%, and the instantaneous lumbar forces (compression and shear) were not markedly high. However, lumbar flexion during the job was close to the full range and was sustained for prolonged periods of time. The pushing forces were also higher than the preferred level and close to the maximum capacity of the workers. The increased lumbar flexion and cumulative demands may explain the high incidence of low back disorders in the punching job of steel workers. The positioning phase of the punching job requires interventions such as worker training and workstation redesign to reduce the risk of WLBD. Fitness to work, job modifications, and training programs can be assessed and designed based on the results.

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Chapter 13 – Comparison of the Lumbar Movement and Lordosis of Nurses and Steel Workers

This Chapter presents an abstract presenting a comparison of the lumbar lordosis and range of motion of the nurses and steel workers. This abstract has been submitted to the World Physical Therapy Congress (Vieira and Kumar, 2006).

13.1. Purpose

Steel workers and nurses have high rates of low back disorders. Data on the lumbar lordosis and range of motion (ROM) of these groups of workers are necessary to assess alterations and treatment. The objective of this study was to quantify the lumbar lordosis and ROM of steel workers and nurses.

13.2. Relevance

Clinical guidelines on the lumbar lordosis and ROM are important to evidence based physical therapy practice.

13.3. Participants

Twenty-three male steel workers and 23 female nurses with no low back disorders resulting in time off work during the previous twelve month participated in the study. The volunteers received explanations about the objectives and procedures and signed a consent form. The mean (SD) age of the steel workers and nurses was 27 (5) and 35 (7) years; the height was 178 (1) and 168 (5) cm; the body mass was 83 (13) and 74 (8) kg, and the body mass index was 26 (4) and 26 (5) kg/m^2, respectively.

13.4. Methods

This study was approved by the University Ethics Committee for Human Studies. The lumbar ROM in the frontal and transversal planes was recorded using electrogoniometers; the lumbar lordosis and ROM in the sagittal plane was recorded using perpendicular markers photogrammetry. Three lumbar ROM recordings were made for flexion, extension, lateral flexion, and rotation, and seven neutral postures were recorded per subject to determine the lumbar lordosis.

Descriptive analysis was performed, including measures of central tendency, and measures of variability. The lordosis and ROM between the groups of workers were compared using one-way ANOVA in SPSS. The homogeneity of the variances was checked using Levene statistic. The significance level was set to 0.05.

13.5. Results

The mean lumbar lordosis was 33° (SD = 15°, 95%CI 30°-35°) for the steel workers and 40° (SD = 13°, 95%CI 38°-43°) for the nurses. The difference on the lordosis between the groups of workers was significant ($F = 22.3$, $p < 0.001$). There were no significant differences between right and left flexion and rotation for both groups. The mean (SD) lumbar ROM of the steel workers and nurses were, respectively, 49° (17°) and 53° (8°) for flexion, 11° (6°) and 15° (9°) for extension, 25° (5°) and 24° (5°) for lateral flexion, and 12° (5°) and 18° (12°) for rotation. In general, the nurses had higher lumbar ROM than the steel workers. The differences between the lumbar ROM of the groups was significant for extension ($F = 8.46$, $p = 0.004$) and rotation ($F = 16.89$, $p < 0.001$).

13.6. Conclusions of Chapter 13

There are differences in the lumbar lordosis and ROM between steel workers and nurses. The differences may be explained by the gender of the workers and is possibly affected by the work activities. Future studies could partition the roles of job adaptations and gender differences on the determination of lumbar lordosis and ROM. In addition, data on workers with low back disorders may be compared with that of the healthy workers presented. The information provided may contribute to evidence based physical therapy practice.

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Chapter 14 – General Discussion and Conclusions

This chapter starts by pointing out how this research relates to the areas of work physical therapy and rehabilitation ergonomics, and thereby to the field of occupational health. It highlights why it is relevant for the development of the areas in Brazil (and in other countries) and suggests a framework for practice based on the results. This is important because my PhD was sponsored by the Brazilian government and I intend to work in academe (research and teaching) and consulting practice (prevention of WMSDs) in Brazil. An extended version of this discussion was published in the journal Disability and Rehabilitation (Vieira, 2006a). After contextualization, the results of the different studies conducted are summarized, discussed, and integrated. Specifically for the nurses, the integrated results were discussed on a paper titled "Nurses have high incidence of low back disorders even when patients are not obese - why and what can be done to reduce risk?" which was submitted to the journal Bariatric Nursing and Surgical Patient Care (Vieira, 2006b).

14.1. Work-Related Musculoskeletal Disorders (WMSDs) In Brazil and a Framework for their Control and Rehabilitation

Twenty five years ago, and again in 2001, prevention, rehabilitation, and equalization of opportunity for the handicapped were defined as the primary goals of the World Program of Action of the World Health Organization (2001). Work physical therapy and rehabilitation ergonomics are instrumental areas for the fulfilment of these goals because the main objectives of both areas are prevention and rehabilitation of work-related musculoskeletal disorders (WMSDs). WMSDs include work-related low back disorders, shoulder and wrist tendinitis, and carpal tunnel syndrome (BLS, 2001). In Brazil, WMSDs are defined as musculoskeletal problems whose frequency, causation, and/or severity are modified by the work, and they are among the most common work-related health problem (Jucá *et al.*, 2005). In 2002, a total of 105,514 work-related "diseases" were recorded by the Brazilian National Institute of Social Insurance (*INSS*). However, these figures are gross under-estimations of the problem because the information includes only the employed workers (N = 75,471,556 in 2002, according to the Brazilian Institute of Geography and Statistics – *IBGE*) covered by the Work

Accidents Insurance (*SAT*) from the *INSS* (n = 22,903,311 in 2002, according to the *IBGE*). Thus, only one-third of the working population was included in the sample, and the number of cases is likely to be higher. Actually, based on a survey (n = 1072 workers) it was estimated that 310,000 workers (6.6% of the working population) suffered WMSDs in 2001 in São Paulo city alone (*Ministério da Saúde*, 2001). Improved surveillance and injury recording systems are needed in Brazil. In addition, occupational health professionals and ergonomists need to work closer to alleviate work-related problems (O'Neill, 2000). This is important and has repercussions for the practice, certification, and regulation of the work physical therapy area. The following framework is derived from this doctoral study and is suggested for practice.

When the work physical demands exceed the workers functional capacity, the risk of WMSDs is significantly increased. The prevention and rehabilitation of WMSDs involve the work physical demands analysis (PDA) and the workers/patients' functional capacity evaluation (FCE). The PDA objectives are to measure the physical effort required by the job including the postures, movements, repetitions, and duration of the work tasks. While the FCE objectives are to determine the functional status of the workers/patients including their range of motion and force capabilities. The FCE information can be used to assess if the subjects are fit for the job both at admission and when returning to work after a WMSD. The study of the patients, workers, and jobs should be systematic and should use the percentages of the normal or maximal available and required motion, posture, force, repetition, and duration (Vieira and Kumar, 2004a,b). The relationship between job requirements and the workers resources was further explored by Kumar (1989 and 1992) including the social and psychological dimensions, in addition to the physical demands of work.

The PDA and FCE should be considered together for optimum outcomes. When a WMSD happens (if not before that) a PDA should be performed. In addition, the injured worker should go trough a FCE. The results can then be used to study the relationship between the injury and the work demands. They should also be used to design the rehabilitation program for the injured worker, and to plan work modifications to prevent similar incidents. After the rehabilitation program is implemented, a second FCE should be performed to evaluate whether the worker (I) has gained or recovered the required physical capacity to perform his/her job; or (II) is fit to return to work; or (III) needs further rehabilitation and its objectives; or (IV) needs to change jobs and what activities would he/she be able to perform.

Similarly, after the job modification plan is put in place, a new PDA should be conducted to assess if (I) the modifications were sufficient to adjust the physical demands and reduce the risk of new cases and re-injuries, or if (II) further interventions are required and what they should be. In the long term this procedure will result in an efficient surveillance system and pro-active management of WMSDs.

Further interaction between the areas of work physical therapy and rehabilitation ergonomics may contribute to improved WMSDs prevention and rehabilitation. The research presented in this thesis is a step in this direction because it used and presented a combined methodology to determine injury rates, to assess the problems and possible improvements based on the workers opinions, and to perform a FCA and a PDA that can be used to evaluate different workers and occupations. The study was performed focusing on low back disorders, but the rational and framework proposed is also applicable to other WMSDs.

14.2. Discussion of the Total Results and Integration of the Studies

The initial work provided the background information used and constituted the foundation of the thesis. The reviews were used to define the risk factors for low back disorders, the rates of injury in the population and in different jobs, and arrive at a definition. It provided knowledge about working postures and movements, including the devices used to measure them. Based on the information gathered, it was possible to delineate the method of normalizing the values by the workers capacity. It also provided information regarding the forces and the guidelines available in the literature. Finally, it provided an overview of the load estimation methods currently in use.

The epidemiological reviews provided the information regarding the jobs with highest rates of WLBD in the worksites analyzed and presented an overview of the dimension of the problem in these worksites. It pointed towards the main problems and risk factors associated with the recorded incidents and guided the choice of the jobs to be further studied with the questionnaire surveys. The questionnaire surveys provided the workers point of view in relation to the risk factors and possible improvements. This information was crucial to determine the interventions that may result in better compliance during a participatory ergonomic intervention.

The questionnaire information also permitted the identification of the critical tasks within the jobs, as well as other factors not directly related to the work itself such as the increased risk among smokers and overweight workers. Patient transfers, turning and repositioning patients in bed, and the punching task were considered the most physically demanding and risky parts of the jobs by the ON, IN, and SW, respectively.

The functional capacity evaluations provided information regarding the physical capabilities of the workers including range of motion, force and related muscle activity. The information was used to normalize the job exertions and to define preferred levels of exertion. There were differences between the forces and EMG during maximum voluntary exertion (MVE), job simulated, and preferred levels for lifting (squat and stoop), pushing, and pulling. There were also differences between the nurses and steel workers' lumbar lordosis ($F = 22.3, p < 0.001$) and ROM for extension ($F = 8.46, p = 0.004$) and rotation ($F = 16.89, p < 0.001$).

The physical demand analysis studies provided information regarding the biomechanical loads on the lumbar spine of the workers during the jobs. The information included the analysis of the lumbar motions and postures, and the estimation of the compression and shear forces, ligament strain, and population capable of performing the jobs and tasks analyzed. The different variables assessed permitted the identification of the different risk factors in the tasks analyzed. For example, if only the lumbar loads were assessed for the steel workers, then the punching job would have been considered as low risk. However, the motion analyses showed that the ranges used were too close to the maximum and that increased flexion was sustained for prolonged periods of time.

Table 14-1 presents selected results from the different studies in an integrated form. The results highlight the relationship between the work physical/biomechanical demands, the mismatch between functional capacity and demands, and the rates of work-related low back disorders (WLBD).

Table 14-1. Integrated results demonstrating the association between high work demands, mismatch between workers capacity and demands, and rates of work-related low back disorders (WLBD): mean (standard deviation).

	Rates of WLBD			Work Physical Load Variables											
Job	Point Prev. of Low Back Pain (%)	Annual Incid. of WLBD (%)	Working -Life Incid. of WLBD (%)	Low Back Discomfort (10-point scale)	Weight Manually Handled (kg)	Perceived Exertion (10-point scale)	% of Max. Effort Used	Peak % of Range of Flexion Used	Average % of Range of Flexion Used	% of Max. Force Used	Instant. Compres. at L5/S1 (N)	Instant. Shear at L5/S1 (N)	% of Pop. without Sufficient Torso Strength	% of Ligament Strain
Ortho Nurses	30	4.3	65	4.6 (3.0)	47 (30)	7 (2)	67 (14)	95 (17)	57 (18)	79 (16)	4754 (437)	487 (40)	37 (9)	14 (5)
ICU Nurses	25	3.1	58	4.0 (2.6)	26 (10)	6 (2)	68 (15)	68 (22)	35 (15)	76 (12)	4048 (447)	454 (53)	27 (8)	15 (2)
CNC Steel Workers	16	5.4	36	5.1 (2.8)	35 (11)	5 (1)	67 (11)	90 (13)	68 (16)	72 (22)	2828 (318)	219 (33)	5 (1)	16 (2)
Welders	27	3.4	55	3.6 (2.6)	21 (6)	5 (1)	68 (17)	NA	NA	NA	NA	NA	NA	NA

NA - not assessed.

The point prevalence of low back pain and working-life incidence of WLBD followed similar trends for the four groups of workers studied (orthopedic nurses, ICU nurses, CNC steel workers, and welders). The annual incidence rate of recorded WLBD also followed similar trends. However, the CNC steel workers had higher rates of recorded WLBD. The annual incidence rates of recorded WLBD were similar to the ones previously reported for other heavy jobs. For example, Klein *et al.* (1984) found the following annual rates of WLBD: 3.3% for nurses, 3.3% for lumbermen, and 2.8% for construction workers. Myers *et al.* (1999) found the following annual rates of WLBD: 3.6% for public workers, 3.1% for recreation and parks' workers, 2.4 for transportation workers, and 0.5% for education workers (a lighter job).

There are several problems with the epidemiology of low back disorders such as diagnostic classification, inadequate surveillance, legal, social, financial and psychological issues/confounders. Data is always approximate (Andersson, 1991). The low back discomfort rates by the end of the shift followed a similar trend to the annual incidence rates of WLBD. The weights handled and perceived exertion trends mirrored each other and followed a similar ranking pattern to the point prevalence of low back pain and working-life incidence of WLBD. The percentage of the maximum effort used rated on Visual Analogue Scales was not sensitive enough to differentiate between the jobs.

Unfortunately, we were not able to evaluate the functional capacity or to quantify the physical demands of the welders because of their busy schedule and lack of management support for completing the study within that worksite. For the orthopedic nurses, ICU nurses, and CNC steel workers, the peak percentages of the flexion range of motion and the percentages of maximum force used mirrored each other and followed similar trends to the point prevalence of low back pain and working-life incidence of WLBD; while the average flexion mirrored the annual incidence rates' sequence.

The instantaneous compression and shear, and the population without sufficient torso strength reflected similar trends and followed the patterns of the point prevalence of low back pain and working-life incidence of WLBD. While the estimated percentage of ligament strain had the same tendencies as the annual incidence rates of recorded WLBD. Finally, the magnitude of the sum of the mean work physical load variables was 5649 for the orthopedic nurses, 4827 for the ICU nurses, and 3410 for the CNC steel workers.

Thus, the sums of the means of the variables were related to the ranking of the point prevalences of low back pain and working-life incidences of WLBD for the jobs.

Environmental, biomechanical, organizational, personal, genetical, psychosocial, psychological, and financial factors all interact and contribute to different extents to the causation of low back injury, disorders, pain, and disability and/or on reporting WLBD. Given the multifactorial nature of these disorders it is not feasible to completely eliminate/prevent them (Kumar, 2001). However, it is possible to improve the control of WLBD, and in order to do so, it is necessary to assess and modify those risk factors that allow for intervention. The results of this research show an association between perceived and actual physical loading at work and WLBD rates corroborating the results of previous studies that have found evidence that awkward postures, heavy physical work, lifting, and forceful movements have causal relationship with WLBD (Bernard, 1997). All jobs evaluated were heavy and had high rates of WLBD in relation to the general population and to other occupations (Bejia *et al.*, 2005; Buckle, 1987; Dueker *et al.*, 1994; Engkvist *et al.*, 1992; Hildebrandt *et al.*, 1996; Jensen, 1987; Josephson *et al.*, 1997; Klein *et al.* 1984; Lagerstrom *et al.*, 1998; Masset and Malchaire, 1994; National Academy of Sciences, 2001; Udo and Yoshinaga, 2001).

WLBD control measures include designing the job for all workers, selecting workers with low risk of developing low back disorders as consequence of specific work demands, and training the workers to perform the tasks as safely as possible (Snook, 1988). Training and education combined with ergonomic interventions, using mechanical lifts, and exercising regularly were found to be effective in reducing WLBD rates (Bos *et al.*, 2006; Garg and Owen, 1994; Li *et al.*, 2004; Skargren and Oberg, 1996). However, worker selection and training alone are not enough to reduce low back disorders in the workplace (Hignett, 1996). Thus, the redesign of highly demanding jobs is fundamental for a successful ergonomic intervention. It has been proposed that designing the jobs' physical demands to fit the workers' functional capacity can potentially reduce up to one-third of work-related back disorders (Snook, 1978). This theoretical proposition was more recently corroborated by Hignett (2001) who reported a successful five-year intervention program resulting in 33% reduction in manual handling incidents, and 36% reduction in days lost due to musculoskeletal disorders.

Despite the long recognition of the need for adjusting the job to reduce the physical demands, a review of 25,291 manual material handling tasks identified the ongoing need of physical workload adjustment (Ciriello *et al.*, 1999). Lifting and lowering comprised approximately 70% of the tasks. Control efforts should be performed to decrease the number and loads of the lifts and lowers, minimizing hand distances, increasing heights of start of the lifts, and decreasing the distances of carries, pushes and pulls (Ciriello *et al.*, 1999). The introduction of new assistive devices results in higher worker compliance to recommendations (prevention programs) than training alone (Daynard *et al.*, 2001). The use of assistive devices, such as mechanical patient lifts, reduces peak spinal loading (e.g. Marras at al., 1999; Silvia *et al.*, 2002). However, mechanically assisted transfers tend to take significantly longer resulting in higher cumulative loads in some cases (Daynard *et al.*, 2001). Each situation and control measure needs to be evaluated separately and in combination to identify the most effective interventions or most efficient combination of control measures. A strategy to control WLBD is to identify the risk factors and intervene to eliminate or alleviate their effects. Effective WLBD control measures depend on adequate surveillance systems, risk assessments, and evidence based intervention measures (Troup *et al.*, 1988).

- The different studies conducted evaluated different variables related with the risk of WLBD in the jobs.
- The different studies conducted lead to corroborating conclusions.
- Each study contributed significantly to clarify different facets of the overall risk of WLBD.
- The combined results provide a much clearer understanding of the problem and of the necessary corrective measures to reduce the risk of WLBD in the jobs evaluated.
- The combined methodology used included epidemiological reviews, questionnaire surveys, qualitative analysis, functional capacity evaluations using direct measures of force, muscle activity, and joint motion, physical demands analyzes using video-recording, photogrammetry, motion analysis, and spinal loads estimations.
- The combined methodology used was useful to identify specific risks to WLBD in the jobs.

14.3. Conclusions

This research studied work-related low back disorders (WLBD) in heavy jobs (nurses and steel workers). It provided background information and a theoretical framework for assessing this problem. The research identified nursing and steel working as occupations with high rates of WLBD. Particularly risky jobs were identified within these occupations. The important role of the workers themselves in identifying the risk factors was demonstrated. Nursing tasks impose significant biomechanical demands on the lumbar spine. Large flexion movements and forces are critical aspects of the transfers requiring most of the nurses' capabilities. For the steel workers, between 95 and 99% of the population would be capable of performing the job based on the biomechanical model estimations and the instantaneous forces were not markedly high. However, the amount of lumbar flexion used was close to the full range; the sustained posture, high flexion movements, and pushing forces were critical aspects of this job and may help to explain the higher incidence of WLBD in these workers. The utility of an analysis taking into account the workers' capabilities was established. The study showed that direct measurements of physical loads are sensitive enough to discriminate different risk factors in the jobs, tasks, and phases. The results support the evaluation of different loading variables such as movements, postures, forces, repetitions, and durations.

This research demonstrated that it is the load as opposed to general occupation title (e.g. nurses, steel workers) that is related to injury rates. Otherwise, both CNC worker and welders (steel workers) and both orthopedic and ICU "nurses" would have the same rates of injury. This information adds to the field by indicating that the studies should focus on the loads as opposed to general occupational titles. It helps to explain why some previous epidemiological studies were not successful in predicting injuries based on occupational titles alone. Hopefully this finding will be taken into consideration in future studies to advance our understanding of the relationship between loading and injury precipitation. Future studies are necessary to establish valid exposure-response relationships to determine definite safe, adequate, and hazardous levels of load exposure. One of the major challenges to this approach is the fact that multiple factors such as movements, postures, forces, repetitions, and durations interact to determine the load. To arrive at a load index it will be necessary to establish adequate multipliers for the load variables. These cut-points need to be developed to establish exposure-response models to estimate how many people are at risk of been affected at different levels of exposure. Only then we will be able to create valid public policies to prevent low back and other musculoskeletal disorders.

Future studies could evaluate prospectively whether the implementation of the recommendations and interventions suggested based on the results of this thesis are effective in reducing the rates of WLBD in these jobs. Thus, the predictive validity of the recommendations (specific training programs, exercise and smoking cessation campaigns, and job modifications) still need to be tested. Despite this, the research was able to provide evidence based recommendations for modifications and training programs to reduce the risk of WLBD in nurses and steel workers. Fitness to work, job modifications, and training programs can be designed and assessed based on the results.

- The combined methodology used to analyze different jobs was demonstrated to be feasible for field use – it was sensitive to identifying specific risk factors for work-related low back disorders and differences between and within the jobs.

- The framework created and the combined methodology tested and presented are significant contributions provided by this research.

- This research demonstrated that it is the load as opposed to general occupation title (e.g. nurses, steel workers) that is related to injury rates.

- This research has added to the scientific literature by demonstrating common ground between ergonomics research and consulting practice, advancing the strategies and methods of work-related low back disorders' control.

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