Firefighter Fitness, Movement Qualities, Occupational Low-Back Loading Demands and Injury Potential

by

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Author's Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

Abstract

Background and Objectives

Low-back overexertion injuries represent a large proportion of fireground "strains, sprains and muscular pains" and are a leading cause of disability and early retirement in firefighters. Given the inherently hazardous and unpredictable nature of many fireground activities, it is often infeasible to implement "task-focused" ergonomic controls and there are limited options to accommodate injured firefighters. Accordingly, effective and practical "worker-focused" injury prevention approaches are needed. Toward this end, four studies were conducted to address the following global thesis objectives:

1) Examine the possible role that firefighters' personal movement strategies could have on their occupational low-back loading demands and injury potential; and
2) Compare the effects of two different exercise approaches on firefighters' occupational low-back loading demands and injury potential.

Study 1: Low-Back Loading Demands during Simulated Firefighting Tasks – Inter-Subject Variation and the Impact of Fatigue and Gender

Background: Non-modifiable fireground duties are considered hazardous for low-back health, but personal movement strategies could modulate low-back loading demands and injury potential. Study objectives were to quantify low-back loading demands during simulated firefighting tasks and to examine the impact of fatigue and gender on the peak loading response.

Methods: Ten men and 10 women performed a battery of laboratory-simulated firefighting tasks before and following repeated bouts of a fatiguing stair-climbing protocol. An EMG-assisted three-dimensional dynamic biomechanical model was used to compute peak L4/L5 joint forces during task performance.

Results: Peak low-back loading demands varied considerably between subjects and tasks, but 70% of all loading variables examined were of greater magnitudes in male subjects and 40% of all loading variables were of lower magnitudes in both males and females after stair-climbing. Some inter-subject variation in low-back loading was attributed to body size differences, but between- and within-subject differences in movement strategies also contributed to low-back loading variability between subjects and over time.

Conclusions: Results of this study suggest that characteristics of individuals, tasks performed, and physical fatigue may influence peak low-back loading demands and injury potential in firefighters. Despite considerable inter-subject variation in the internal low-back loading response to fixed external task and environmental constraints, opportunities to attenuate low-back loading demands through movement behaviour adaptations alone may be limited to only a subset of fireground activities.

Study 2: Ankle Immobilization alters Lifting Kinematics and Kinetics – Occupational Low-Back Loading Demands and Potential for Injury

Background: Firefighters with lingering lower extremity functional impairments could be forced to move in ways that increase their potential for sustaining occupational low-back lifting injuries. The study objective was to examine the impact of unilateral ankle immobilization on lifting kinematics and kinetics.

Methods: With and without their right ankle immobilized, 10 male volunteers performed laboratory-simulated occupational lifting tasks. Together with force platform data, three-dimensional kinematics of the lumbar spine, pelvis, and lower extremities were collected, and a three-dimensional dynamic biomechanical model was used to calculate peak low-back compression and shear loading demands.

Results: In comparison to the unaffected conditions, ankle immobilization resulted in less knee (*p*-values between 0.0004 and 0.0697) and greater lumbar spine (*p*-values between 0.0006 and 0.3491)

sagittal motion when lifting. Associated with this compensatory movement strategy were greater L4/L5 anterior/posterior reaction shear forces (p-values between 0.0009 and 0.2450). However, in a few cases where individual compensatory movement strategies differed from the "group" response (i.e., subjects increased their sagittal knee and hip motion on the affected side), peak L4/L5 joint compressive loads increased while the peak L4/L5 anterior-posterior shear did not change.

Conclusions: Distal lower extremity joint dysfunction can alter the way in which individuals move and load their low-backs when lifting. The specific ways in which individuals compensate for personal movement constraints could alter the potential site and mechanism of occupational low-back injury.

Study 3: FMS™ Scores and Occupational Low-Back Loading Demands – Whole-Body Movement Screening as an Ergonomic Tool?

Background: Results of Study 1 suggested that a whole-body movement screen could be used to identify personal characteristics that constrain movement behaviour in ways that impact occupational low-back loading demands and injury potential. The purpose of this study was to examine if Functional Movement Screen™ (FMS) scores could be used to project the low-back loading response to lifting.

Methods: Sagittally symmetric and asymmetric laboratory-based lifting tasks were performed by 15 firefighters who scored greater than 14 on the FMS (high-scorers) and 15 size-matched low-scorers (FMS < 14). A three-dimensional dynamic biomechanical model was used to calculate low-back loading demands, and lumbar spine posture was recorded when peak low-back compression was imposed.

Results: Regardless of the task performed, there were no differences in peak L4/L5 joint compression ($p \geq 0.4157$), anterior/posterior reaction shear ($p \geq 0.5645$), or medial/lateral reaction shear ($p \geq 0.2581$) loading demands between high- and low-scorers. At the instant when peak compression force was detected, lumbar spine deviation was not different between high- and low-scorers about the lateral bend ($p \geq 0.4215$), axial twist ($p \geq 0.2734$), or flexion/extension ($p \geq 0.1354$) axes.

Conclusions: Using the previously established musculoskeletal injury prediction threshold value of 14, the composite FMS score did not project the low-back loading response to lifting. Future attempts to modify or reinterpret FMS scoring are warranted given that several previous studies have revealed links between composite FMS scores and musculoskeletal complaints.

Study 4: Movement- vs. Fitness-Centric Exercise – Firefighter Fitness, Whole-Body Movement Qualities, and Occupational Low-Back Loading Outcomes

Background: The impact of exercise on firefighter job performance and cardiorespiratory fitness has been studied extensively, but its effect on musculoskeletal loading remains less understood. The aim of this study was to compare various physical fitness, general movement quality, and low-back loading outcomes between groups of firefighters who completed fitness- or movement-centric exercise.

Methods: Fifty-four firefighters participated and were assigned to a control (CON), fitness-centric exercise (FIT), or movement-centric exercise (MOV) group. Before and after 12 weeks of exercise, subjects performed a physical fitness test battery, the Functional Movement Screen™ (FMS), and laboratory-simulated firefighting tasks during which low-back loading demands were quantified.

Results: FIT and MOV subjects exhibited statistically significant improvements in nearly all measures of physical fitness (i.e., body composition, cardiorespiratory capacity, muscular strength, power, endurance, and flexibility), but FMS scores and occupational low-back loading demands were not impacted in a consistent way across individuals.

Conclusions: Improving physical fitness can enhance job performance and prevent cardiac events in firefighters, but it was not clear that 12 weeks of exercise would alter their occupational low-back loading demands. Given variability in individual responses, the short study duration, and limited number and nature of tasks examined, more research incorporating alternative biomechanical and statistical analyses is needed to better understand how individuals adapt to chronic exercise and what impact these adaptations have on occupational movement behaviours, low-back loading demands, and low-back loading capacity.

Summary and Conclusions

Results confirmed that fireground activities are potentially hazardous for low-back health, as simulated occupational low-back loading demands routinely exceeded recommended exposure limits in the studies performed. However, results also indicated that personal movement strategies – possibly influenced by body size, preference, gender, physical fatigue, or distal lower extremity joint dysfunction – could alter occupational low-back loading demands and injury potential. It could not be concluded that occupational low-back loading demands and injury potential would be consistently affected by short-term improvements in physical fitness, nor could the low-back loading response to lifting be projected by scoring above or below 14 on the Functional Movement Screen™. Future research is warranted to examine the low-back loading demands associated with performing non-fireground duties, as opportunities may exist to implement ergonomic strategies to control cumulative low-back loading exposures. Particular attention should be paid to the exercise and training practices of firefighters, as musculoskeletal injuries sustained during these activities are potentially avoidable and could reduce the capacity of the musculoskeletal system to withstand demands imposed during non-modifiable fireground operations.

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Not only did my friend and mentor, Jack Callaghan, inspire and motivate me to aim high, work hard, and see things through, but he also afforded me numerous opportunities to grow personally and professionally over many years working together. He provided an outstanding training environment and seemingly limitless resources to pursue our shared research interests, and he effectively taught me almost everything I know about biomechanics. I am extremely fortunate to have been trained by a respected, rigorous, and highly productive scientist, and none of this work would have been possible without his constant support, encouragement, and friendship.

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CHAPTER 1

General Introduction

CHAPTER 1

General Introduction

Firefighters perform inherently hazardous work and are more likely to sustain job-related musculoskeletal injuries than most other workers (Maguire et al. 2005). In addition to the obvious personal costs related to suffering line-of-duty injuries (e.g., chronic pain, suffering, and disability), firefighter injuries are very expensive in economic terms, as economic costs associated with disability payments, medical treatments, duty restrictions, lost work and training time, and attrition could be as high as $7.8 billion in the United States alone (NIST 2005). Accordingly, there is considerable motivation to reduce the number and impact of firefighter injuries.

Over the past 30 years, the frequency of musculoskeletal injuries reported during fireground operations has steadily increased (Figure 1.1). The largest proportion of these injuries is attributed to "overexertion" (Karter and Molis 2010), and over 40% of firefighter overexertion injuries affect the low-back (Walton et al. 2003). On the fireground, firefighters can be exposed to a number of physical risk factors for low-back pain development, including: heavy physical work; forceful exertions; and awkward postures (NIOSH 1997). These physical risk factors are easily recognized at fire scenes, but it is often infeasible to implement "task-focused" engineering or administrative controls due to the unpredictable and non-modifiable nature of many fireground activities. In such cases, effective and practical "worker-focused" low-back injury prevention approaches are needed, especially for uniformed personnel whose low-back loading capacity may have diminished over time.

Knowledge derived from biomechanical tissue testing and biologically-assisted musculoskeletal models suggest that occupational low-back loading demands, capacity, and injury potential can be highly sensitive to personal movement strategies (McGill 1997; McGill 2004; McGill 2009). Thus, interventions designed to impact movement behaviour could conceivably be implemented when task and environmental characteristics cannot be controlled. In fact, even when jobs can be re-designed based on fundamental principles of occupational biomechanics and ergonomics, opportunities to prevent low-back injuries could be ineffective if tissue-damaging movement behaviours are unaffected by the intervention (McGill 2009). In response to an ergonomic intervention, for example, workers may adapt their movement behaviour and mitigate the intended effect (Faber et al. 2007). Congruent with the message championed by McGill (2009), a *Movement Matters!* perspective is advocated in this thesis.

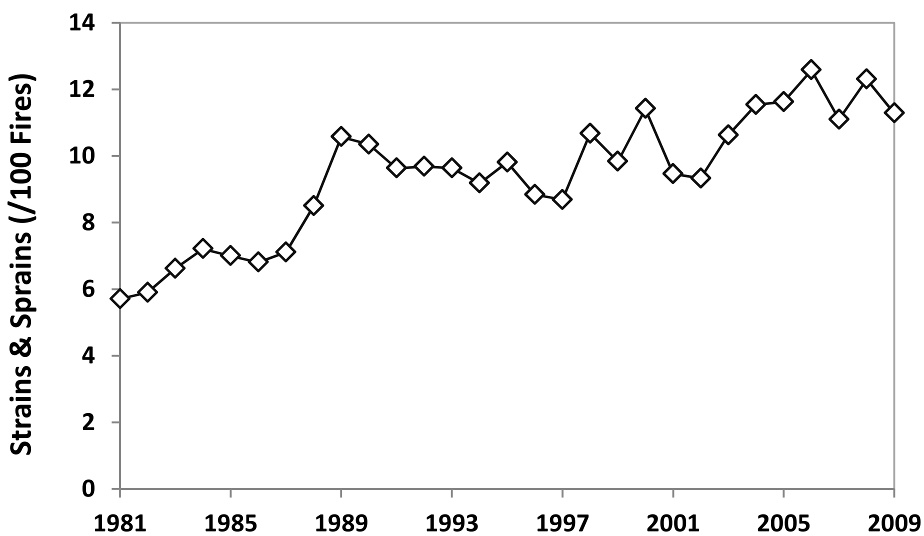

Figure 1.1. Annual survey data published by the National Fire Protection Association (NFPA)[1] indicate that the frequency of musculoskeletal injuries ("strains and sprains") suffered during fireground operations has steadily increased between 1981 and 2009. Data included in this figure were acquired from the NFPA website (last accessed on February 9, 2012): http://www.nfpa.org.

[1] Each year, the NFPA surveys a sample of fire departments in the United States to estimate the number of firefighter injuries. In the most recent survey (2009), over 2,700 departments responded. National projections are produced by weighting sample data as a proportion of the total United States population that is accounted for by the community sizes serviced (by responding departments). Confidence intervals calculated in 2009 indicate that the number of total injuries is within 6.3% of the estimate (Karter and Molis 2010).

1.1. Global Thesis and Personal Objectives

Global Thesis Objectives

From a *Movement Matters!* perspective, there were two global objectives of this thesis:

Global Objective 1: Examine the possible role that firefighters' personal movement strategies could have on their occupational low-back loading demands and injury potential

Together with task goals and environmental make-up, structural and functional attributes of individual performers may well interact to influence movement behaviour (Davids et al. 2003). Owing to the multitude of modifiable and non-modifiable movement system components and potential for interactions within and among task, environmental, and individual movement constraints, attempts to objectively identify the number and relative importance of personal characteristics that affect movement behaviour at work may prove difficult. However, it was hypothesized that there are "gross" observable personal qualities that could (consciously or subconsciously) promote potentially injurious movement strategies at work.

Global Objective 2: Compare the effects of two different exercise approaches on firefighters' occupational low-back loading demands and injury potential

Most traditional firefighter exercise programs are "fitness-centric" in that they are designed with the primary objective being to improve firefighters' muscular strength, power, endurance, flexibility, and cardiorespiratory efficiency. While there is evidence to suggest that fitness-exercise does improve firefighters' job performance capabilities and cardiorespiratory functioning, it is unclear if or how occupational low-back loading demands and potential for line-of-duty injuries are affected. In fact, recent research is interpreted by some to suggest that "movement-centric" exercise is perhaps better suited to reduce firefighter occupational low-back injury potential. The movement-focused approach aims to induce stable "joint-sparing" movement behaviour adaptations believed to transfer more directly to physical activities of daily living. To achieve this, movement-centric trainees are urged to attend to the *way* their bodies move when exercising (i.e., internal focus of attention) rather than on the *outcomes* of their exercise movements (i.e., external focus of attention). Focus on exercise outcomes (e.g., quantities of weight lifted, repetitions performed, etc.) is an intrinsic feature of fitness-centric programs.

Personal Objective

The overarching personal goal of completing this thesis was to conduct studies that could be used to lay the groundwork for developing sustainable exercise-based occupational and athletic performance enhancement and musculoskeletal injury prevention strategies. Studies were conceived to address the global thesis objectives, and in so doing, address a subset of questions directly linked to the potential for exercise-based interventions to attenuate occupational low-back loading demands.

1.2. Thesis Lay-Out

Following a literature review that was undertaken to define the scope of the problem and to outline and rationalize salient aspects of the general experimental approaches taken (Chapter 2), four studies are presented in Chapters 3 to 6 that addressed specific research questions related to the global objectives. In Chapter 7, general discussion points, future research recommendations, and thesis conclusions are provided. Supplementary materials are appended (Appendices I to V).

1.3. Specific Research Questions and Hypotheses Tested

Four laboratory-based studies were carried out to test specific hypotheses related to the global thesis objectives:

Study 1: Low-Back Loading Demands during Simulated Firefighting Tasks – Inter-Subject Variation and the Impact of Fatigue and Gender

Non-modifiable fireground operations are deemed inherently hazardous for low-back health. However, it is possible that some individuals move their bodies and activate their trunk muscles in ways that increase (or decrease) low-back loading demands and injury potential. In Chapter 3, a biomechanical model that is sensitive to inter- and intra-individual variability in movement and trunk muscle activation patterns was used to quantify low-back loading during laboratory-simulated firefighting tasks. In addition to documenting the inter-subject variation in low-back loading, the impact of fatigue and gender on low-back loading were also examined. The hypotheses tested were that considerable inter-individual variation in low-back loading demands exists despite fixed external task and environmental constraints, loading demands are different between men and women, and that fatigue alters the loading reponse. If some individuals are able to meet physically demanding firefighter task objectives without exceeding recommended exposure limits, it may be possible to devise interventions for individuals who employ movement strategies that make them more susceptible to suffering fireground injuries.

Study 2: Ankle Immobilization alters Lifting Kinematics and Kinetics – Occupational Low-Back Loading Demands and Potential for Injury

Firefighters with lingering distal lower extremity functional impairments (e.g., due to non-rehabilitated injuries) may be forced to move in ways that increase their potential for sustaining occupational low-back lifting injuries. In Chapter 4, the effect of unilateral ankle joint immobilization on the kinematics and kinetics of lifting was examined. The hypothesis tested was that ankle immobilization would elicit compensatory movement strategies and alter low-back loading demands and injury potential change as a consequence. The possibility that distal lower extremity joint dysfunction could impact low-back loading demands and injury potential during lifting motivated the decision to examine the relationship between whole-body movement screen scores and low-back loading demands and injury potential during lifting.

Study 3: FMS™ Scores and Occupational Low-Back Loading Demands – Whole-Body Movement Screening as an Ergonomic Tool?

The Functional Movement Screen™ (FMS) is envisaged as a tool to identify personal movement constraints (e.g., deficits in joint mobility and neuromuscular control) that promote potentially injurious athletic or occupational movement strategies. Scoring below 14 on the FMS has been linked with the reporting of musculoskeletal complaints in firefighters, and a "corrective" exercise framework exists to improve FMS scores. In Chapter 5, the low-back loading demands in lifting were compared between size-matched firefighters who scored greater or less than 14 on the FMS. The hypothesis tested was that peak low-back loading responses would differ between high- and low-scorers. Outcomes could be used in the decision to include the FMS in worker-focused low-back injury prevention efforts.

Study 4: Movement- vs. Fitness-Centric Exercise – Firefighter Fitness, Whole-Body Movement Qualities, and Occupational Low-Back Loading Outcomes

Fitness-centric exercise approaches are designed with a primary focus of improving measures of physical fitness (e.g., muscular strength, power, endurance, flexibility, and cardiorespiratory efficiency). The primary objective of movement-centric exercise is to elicit stable and transferable "joint-sparing" movement behaviour adaptations. In Chapter 6, physical fitness, FMS, and peak low-back loading outcomes were compared between groups of firefighters who completed 12 weeks of movement- or fitness-centric exercise. The hypothesis tested was that between-group differences in exercise outcomes can be used to justify one exercise-based low-back injury prevention approach over the other. Outcomes of this study could be used to inform exercise-based low-back injury prevention programs for firefighters.

CHAPTER 2

Review of Literature

CHAPTER 2

Review of Literature

2.1. Low-Back Injuries in Firefighters

According to an annual survey-based estimates published by the National Fire Protection Association (Karter and Molis 2010), musculoskeletal strains and sprains account for about half of the 80,000 on-duty injuries sustained by firefighters each year in the United States. It was estimated that over 15,000 musculoskeletal injuries occurred during fireground activities (e.g., set-up, extinguishment, ventilation, and overhaul at fire scenes) in 2009 and the largest proportion (25%) of these injuries was attributed to overexertion (Karter and Molis 2010). Workers' compensation records indicate that nearly half of all firefighter overexertion injuries are related to lifting and over 40% of firefighter overexertion injuries affect the low-back (Walton et al. 2003). Work-related low-back injuries have been identified as the most (Reichard and Jackson 2010) or second most (Poplin et al. 2011) frequently reported of all firefighter musculoskeletal injuries and account for almost half of all line-of-duty injury-related retirements in the United States each year (IAFF 2000). The high disability rates associated with low-back injuries could be related to the fact that few modified work options are available to injured firefighters given the physically demanding and often unpredictable nature of their work.

Firefighter injuries are very expensive in economic terms. In a study issued by the National Institute of Standards and Technology (NIST 2005), it was estimated that total (indirect plus direct) annual costs of "…addressing firefighter injuries and efforts to prevent them…" could be as high as $7.8 billion in the United States alone. Walton et al. (2003) reported that the workers' compensation costs associated with firefighter overexertion injuries (average of $10,000 per claim between 1992 and 1999) are 89% more costly than are injuries attributed to other causes (e.g., slips, trips, and falls). Moreover, the authors found that musculoskeletal injuries are 80% more costly than all other injury outcomes (e.g., wounds and fractures). Though economic costs associated specifically with firefighter low-back injuries were not reported by Walton et al. (2003), work-related low-back disorders tend to be more expensive than other work-related musculoskeletal disorders (Dempsey and Hashemi 1999; Dunning et al. 2010) and there is no reason to expect that this would be any different in the fire service. In fact, low-back injuries could be even more costly in the fire service because of the limited opportunities available to modify various critical and essential firefighter duties.

Given the significant personal and societal costs associated with low-back injuries in firefighters, there is considerable motivation to develop effective and practical low-back injury prevention programs. Moreover, since the frequency of fireground musculoskeletal injuries continues to increase (Chapter 1, Figure 1.1), there is particular interest in devising interventions that can reduce the number and severity of fireground injuries. Toward this end, the research in this thesis was conducted to better understand how personal characteristics could alter the low-back loading response to simulated firefighter task demands. Armed with this knowledge, "worker-focused" interventions can be conceived and tested in cases where "task-focused" ergonomic controls (e.g., job modification) cannot practically or possibly be implemented.

2.2. Occupational Low-Back Injury Framework

If progress is to be made in reducing the number of fireground low-back injuries, it is important to establish a common research framework in order to: identify salient research questions; employ appropriate research methodologies; organize and synthesize research findings; and provide a basis for practical application. In this thesis, the musculoskeletal "strains, sprains, and muscular pains" reported by firefighters are considered to be "injuries", and a recommended framework for studying occupational low-back injuries is adopted (McGill 1997).

According to McGill (1997), low-back injuries are defined as *low-back tissue damage resulting from mechanical overload*. There are two ways in which low-back tissues can be overloaded in a mechanical sense. In the first scenario, tissue damage occurs when a single applied load exceeds the failure load tolerance of the tissue in question. The aforementioned scenario is referred to as an acute or *overexertion* injury mechanism and is depicted in Figure 2.1. The second way in which tissues can be overloaded mechanically is through the repeated or sustained application of loads that are initially of sub-failure load tolerance magnitudes. Chronic or *overuse* injury mechanisms describe tissue-damaging scenarios wherein the ability of tissues to tolerate load decreases in response to repeated (Figure 2.2a) or prolonged static loading (Figure 2.2b).

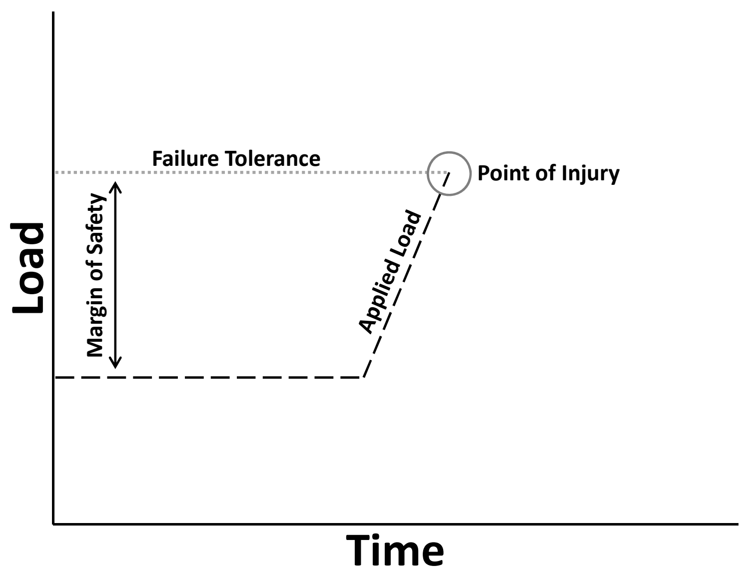

Figure 2.1. A model to describe an acute ("overexertion") injury mechanism (adapted from McGill 1997). Tissue damage (injury) results when the magnitude of a one-time load application exceeds the failure load tolerance of the tissue.

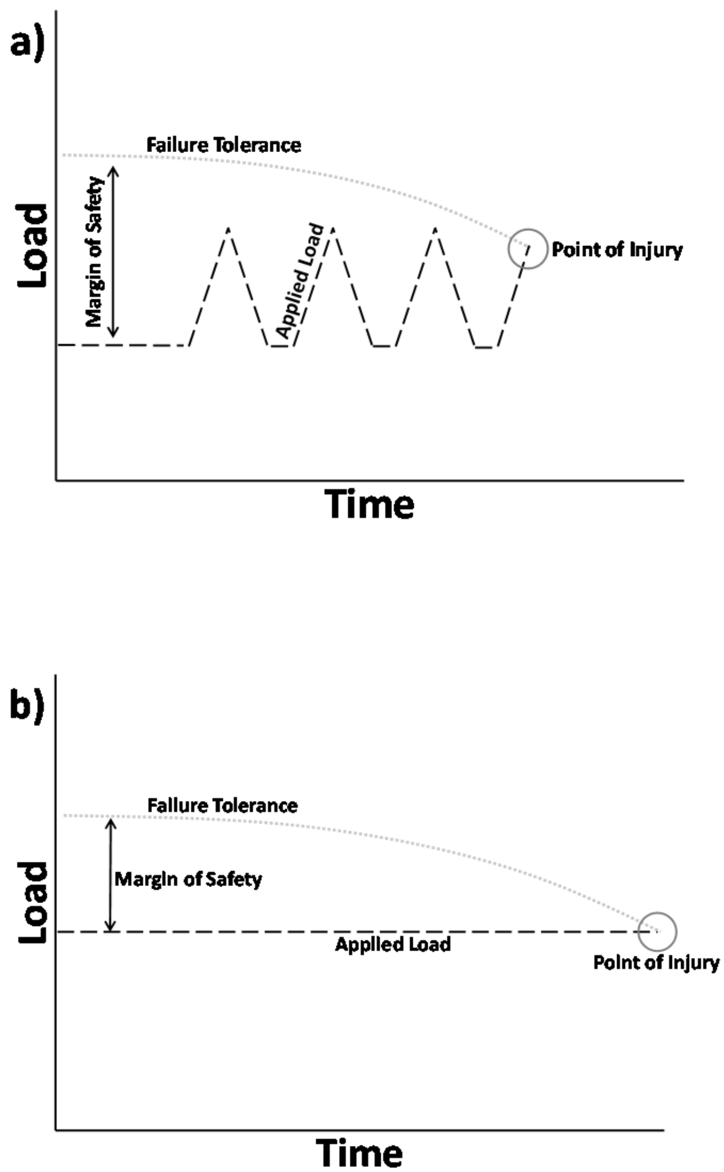

Figure 2.2. A model to describe chronic ("overuse") injury mechanisms (adapted from McGill 1997). Repeated (a) or sustained (b) load application outpaces tissue recovery processes, leading to reduced failure load tolerance. Again, tissue damage (injury) results when the applied load exceeds the failure load tolerance.

Although straightforward in principle, distinguishing between work-related overexertion and overuse low-back injuries is difficult in practice. Clearly, overexertion low-back injury mechanisms can explain tissue-damaging processes that result from accidental workplace events such as slips, trips, and falls. In such instances, applied tissue loads far exceed habitual exposure levels and are thus damaged. However, as McGill (1997) cautioned, discretion must be exercised if aiming to associate the occurrence of low-back injuries with specific occupational "events" unless consideration is given to the short- and long-term low-back loading history. For instance, it is possible to attribute injuries to overexertion causes when the mechanism of low-back injury may be better understood using an overuse injury model; either description could suffice depending on the time frame considered (Figure 2.3).

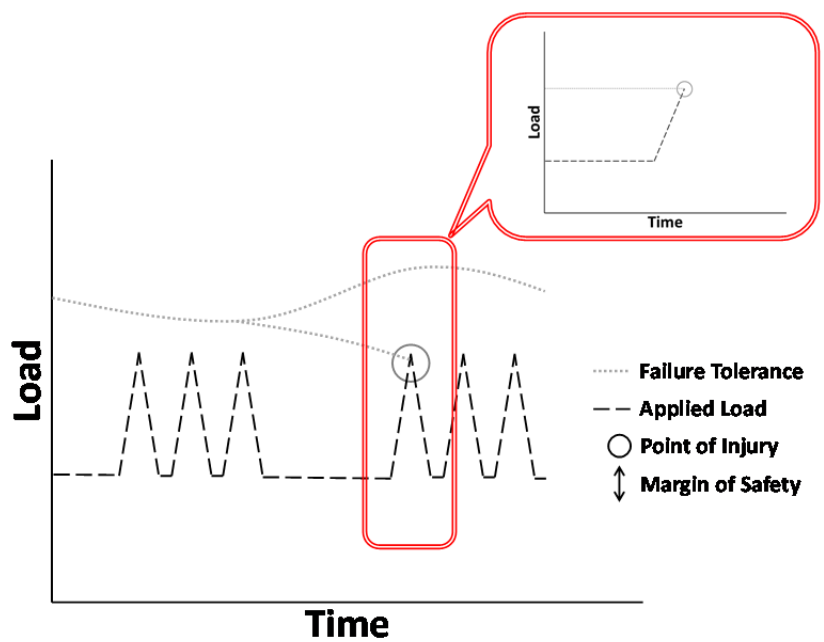

Figure 2.3. Time-varying tissue failure load tolerance. If adequate rest and recovery is afforded to tissues, failure load tolerance can increase. If rest is inadequate to permit tissue recovery and regeneration, failure tolerance decreases and the potential for injury can be increased as a result. Also demonstrated here is that consideration must be given to tissue loading history when attempting to attribute injuries to overexertion or overuse causes. If only a short time frame is considered (area enclosed by double-bordered call-out box), it is possible to attribute injuries to overexertion causes when the mechanism of tissue damage may be better understood using an overuse injury model.

Evidently, an important factor that must be taken into account when exploring potential causal links between work activities and low-back injuries is that tissues adapt (positively or negatively) in response to their mechanical environment (Cowin 1999; Taber 1995). Indeed, adaptive responses to mechanical loading are well documented for dense connective tissues (i.e., bone, cartilage, tendons, and ligaments) (Doschak and Zernicke 2005; Reeves 2006; Ruff et al. 2006; Sommerfeldt and Rubin 2001; Wong and Carter 2003) and skeletal muscles (Baar et al. 2006; Burkholder 2007; Huijing and Jaspers 2005). Appreciating the time-varying nature of tissue failure load tolerance can avert one from adopting the over-simplified notion that attenuating the magnitude of applied tissue loads will always prevent low-back injuries. In fact, epidemiological studies suggest that a *U*- or *J*-shaped function might better describe the relationship between applied tissue load magnitude and injury potential (Magora 1972; Videman et al. 1990) (Figure 2.4). Indeed, there is much experimental evidence to support the contention that tissue failure load tolerance can decrease under conditions of chronic under-loading (Baar et al. 2006; Burkholder 2007; Doschak and Zernicke 2005; Huijing and Jaspers 2005; Reeves 2006; Ruff et al. 2006; Sommerfeldt and Rubin 2001; Wong and Carter 2003), and thus some occupations (e.g., sedentary office jobs) might actually lead to reductions in the load-bearing capacity and health of musculoskeletal tissues due to insufficient mechanical loading at work (Straker and Mathiassen 2009). Conversely, overuse can lead to a progressive reduction in failure load tolerance when work-to-rest ratios render conditions inadequate for tissue recovery and regenerative processes to operate (Figure 2.5). Therefore, when exploring potential causal links between occupational activities and low-back injuries, it must be appreciated that tissue failure load tolerance is not fixed and that the ability of low-back tissues to withstand applied loads without sustaining damage is ultimately dependent on the time-varying mechanical properties of constituent tissues (material properties), the configuration or architectural arrangement of constituent tissues (structural properties), and characteristics of the applied tissue loads.

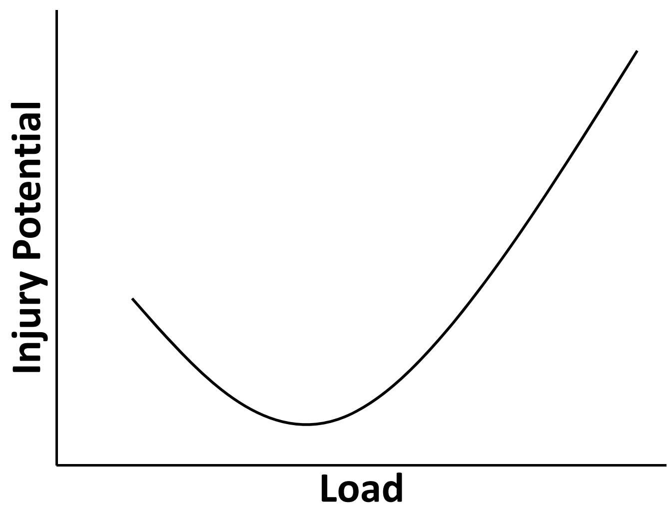

Figure 2.4. Hypothetical relationship between injury potential and the magnitude of applied load; too little or too much load can increase the potential for tissue damage.

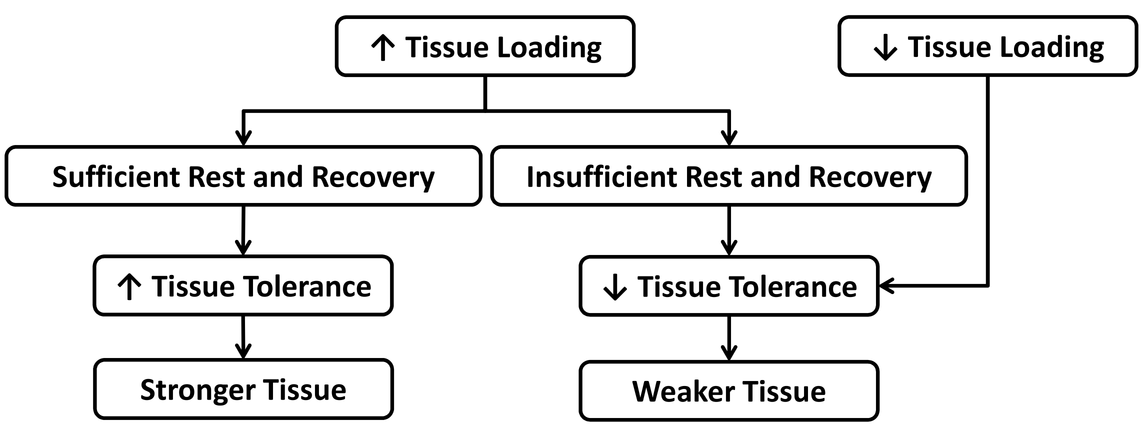

Figure 2.5. Relationship between tissue loading, rest, recovery, and failure load tolerance. Figure adapted from Williams (1993).

Against this backdrop, future success in preventing low-back overexertion injuries on the fireground will most likely be achieved by appreciating that such injuries are influenced, in part, by the amount (frequency, duration, magnitude) and type (mode) of previous occupational low-back loading (e.g., during training and maintenance duties) and non-occupational low-back loading (e.g., during activities of daily living). Future research recommendations are discussed in Chapter 7 which emphasize the need to examine the potential impact of non-fireground low-back loading exposures on firefighter low-back injury reporting. Nevertheless, controlling acute ("peak") low-back loading exposures on the fireground will remain a priority, as success in this regard could also help firefighters maintain their capacity to meet the low-back loading demands associated with all occupational duties.

2.3. Occupational Low-Back Loading Demands

What can be discerned from the low-back injury framework above is that efforts to prevent low-back injuries in firefighters could benefit from knowledge of the low-back loading response to firefighter task demands. Although biomechanical models have been used in previous research to estimate low-back loading demands during manual handling duties that could be carried out by firefighters (Cooper and Ghassemieh 2007; Lavender et al. 2000; Lett and McGill 2006; Lusa et al. 1991), no previous attempts have been made to quantify low-back loading during the performance of fireground-specific tasks (e.g., forcible entry, ceiling breach, hose advance, etc.). Unique constraints imposed by the tasks performed and personal protective equipment worn on the fireground make difficult the extrapolation of previous research. Hence, fireground tasks are usually described qualitatively as being "physically demanding", "heavy", "arduous", and "strenuous" without having corresponding quantitative low-back loading data to incorporate in low-back injury risk assessments. Future success in preventing firefighter low-back injuries could thus be hindered, in part, by the lack of quantitative information available.

For at least three reasons, firefighters likely encounter some of their highest work-related peak low-back loading exposures during fireground operations. First, as highlighted previously in Section 2.1, low-back “overexertion” injuries are frequently reported on the fireground (Walton et al. 2003), suggesting the peak low-back loading levels are of relatively high magnitudes. Second, on the fireground, firefighters are exposed to several of the physical risk factors found by National Institute for Occupational Safety and Health (NIOSH 1997) to be causally associated with the development of low-back disorders (e.g., heavy physical work, forceful exertions, and awkward postures). Third, based on the collective results of a number of studies (Coca et al. 2008; Coca et al. 2010; Granata and Orishimo 2001; Huck 1991; Park et al. 2010; Punakallio et al. 2003; Sobeih et al. 2006; Southard and Mirka 2007), the personal protective equipment worn during fireground operations (e.g., self-contained breathing apparatus and turnout gear) could result in elevated low-back loading levels due to the: mass of the equipment itself; potential need for increased levels of trunk muscle (co-)activation to maintain stable lumbar spine behaviour when the equipment is worn; and possibly because additional load carriage and joint mobility restrictions could promote whole-body movement and postural control strategies that lead to elevated low-back loading levels. Previous studies that have investigated the low-back loading response to firefighter task demands did not incorporate the potential effect of added load carriage (Lett and McGill 2006) and used biomechanical models that were insensitive to potential inter- and intra-individual variations in trunk muscle activation patterns on low-back load magnitude (Lusa et al. 1991; Lavender et al. 2000). In addressing specific hypotheses relevant to the global thesis objective, low-back loading estimates were generated using dynamic three-dimensional biomechanical models that accounted for inter- and intra-individual differences in movement strategies.

2.4. Occupational Low-Back Loading Capacity

It is not currently possible to determine the failure load tolerance of all low-back tissues in individuals. Thus, data derived from *in vitro* mechanical cadaveric tissue testing and/or biomechanical models are used to approximate, most frequently, the compression and shear load tolerance of lower-lumbar intervertebral joint motion segments. In low-back overexertion injury risk assessments, cut-points published by the National Institute for Occupational Safety and Health (NIOSH 1981; Waters et al. 1993) are often incorporated. According to NIOSH recommendations, peak low-back compression forces below an “action limit” of 3.4 kN are considered to be of nominal risk for most workers, whereas peak low-back compression forces above a “maximum permissible limit” of 6.4 kN are deemed hazardous for the majority of workers. However, it should be emphasized that, due to the variability in data and methods used to establish NIOSH limits, low-back overexertion injury risk could be over- or under-estimated in individuals if basing assessments on NIOSH cut-points. Factors such as age, gender, posture, and loading history can influence low-back compression failure tolerance (Genaidy et al. 1993; Jäger and Luttman 1991), but NIOSH limits are based on aggregations of variable cadaveric data sources. It can be discerned from the predictive equations derived by Genaidy et al. (1993) and Jäger and Luttman (1991) that NIOSH limits may not protect older, female, smaller, and previously injured or sedentary workers, but may be conservative if used to assess risk in large young male workers who are physically active and without a low-back injury history. Nevertheless, though it was not the purpose of this thesis to assess risk explicitly, absolute values of peak low-back compression loading estimates in Chapter 3 are compared to NIOSH limits to aid in interpreting the results.

Peak low-back shear tolerance limits are less commonly incorporated in assessments of occupational low-back overexertion injury risk, probably because low-back shear load estimates are highly sensitive to biomechanical modeling assumptions (Dieën and Looze 1999). However, robust associations exist between occupational peak low-back shear forces and low-back pain history (McGill et al. 1998; Norman et al. 1998), and it was hence also decided to relate estimates of peak low-back shear loading in this thesis to recommended exposure limits. Based on a more detailed analyses of the occupational low-back loading and pain-reporting data reported by Norman et al. (1998), McGill et al. (1998) recommended a peak low-back (reaction/joint) shear load "action limit" of 0.5 kN and a "maximum permissible limit" of 1 kN.

There is a growing body of literature indicating that low-back pain and injury reporting is related to measures of occupational cumulative low-back loading exposures (Jäger et al. 2000; Kumar 1990; Norman et al. 1998; Seidler et al. 2001; Seidler et al. 2009; Seidler et al. 2011; Stuebbe et al. 2002), and the overuse injury model presented in Section 2.2 can be used to produce hypotheses regarding underlying causal mechanisms (McGill 1997). As conceded in Section 2.2 and discussed again in Chapter 7, it is possible that fireground low-back overexertion injury potential would be better understood using an overuse injury model. Together with additional information regarding the frequency and duration of fireground activities (estimated using work sampling techniques), future research could use the data in this thesis to approximate cumulative low-back loading exposures on the fireground, and estimates could be interpreted with respect to dose-response relationships reported in epidemiological studies that linked occupational cumulative low-back loading and injury reporting (e.g., Seidler et al. 2009; Seidler et al. 2011).

2.5. ***Movement Matters!***

Advanced low-back biomechanical models incorporating measures of trunk muscle activation, three-dimensional whole-body kinematics and dynamic external contact forces demonstrate that low-back loading patterns are highly sensitive to personal movement strategies (McGill 2004). The number of available movement strategies to meet motor task objectives is constrained by characteristics of tasks, environments, and individual performers (Newell 1986). Indeed, lifting kinematics and associated low-back loading demands have been shown to vary as a function of task and environmental constraints (Davis and Marras 2000b; Faber et al. 2008; Gallagher et al. 2001; Hoozemans et al. 2008; Ning and Mirka 2010; Splittstoesser et al. 2007; Wrigley et al. 2006). And, personal characteristics such as muscular strength, flexibility, obesity, personality, gender, experience, and injury history can also influence how individuals coordinate and control their movements (and load their low-backs) when lifting (Bartlett et al. 2007a; Carregaro and Gil Coury 2009; Li and Zhang 2009; Marras et al. 2000a; Marras et al. 2001; Marras et al. 2004; Marras et al. 2006; Shin et al. 2004; Xu et al. 2008). Therefore, it is posited that in situations where characteristics of work tasks and environments cannot be controlled through ergonomic interventions (e.g., during fireground operations), identifying personal characteristics that could promote potentially injurious low-back loading patterns can assist in designing worker-focused interventions to control occupational low-back loading exposures.

McGill (2009) has persuasively argued that even if work tasks and environments can be modified to “fit work to workers”, progress towards preventing occupational low-back injuries can still be limited if tissue-damaging movement strategies are left unchecked. For example, he argued that individuals who may have the option to rotate primarily about lower extremity joints when lifting, but elect instead to rotate primarily about the lumbar spine, could ultimately damage passive low-back tissues (i.e., intervertebral discs and ligaments) if tissue-damaging movement behaviour persists. Of course, his argument does not suggest that job (re-)design approaches are not preferred or are ineffective. Rather, the argument could be interpreted to suggest that attention should be paid to the impact that interventions have on movement behaviour and associated internal tissue loading patterns. It is particularly important to appreciate that modifying characteristics of work tasks or environments does not necessarily ensure that stable spine “load-sparing” movement behaviour will result following ergonomic interventions. Specifically, individuals have been shown to negate the desired effects of ergonomic interventions by adapting their movement behaviour in response to task modification (Faber et al. 2007), and considerable inter- and intra-individual variations in movement strategies and associated low-back loading patterns are routinely reported (Dieën et al. 2001; Granata et al. 1999).

An important yet sometimes overlooked aspect of McGill's *Movement Matters!* perspective (McGill 2009) is that occupational low-back loading demands and capacity can be simultaneously affected by personal movement strategies. If the lumbar spine approaches full flexion during a sagittal lifting exertion, for example, the effective moment arm, length, and line-of-action of trunk extensor muscles change (Dieën and Looze 1999), posterior passive tissues strain (Adams et al. 1994), and the low-back compression and shear load "margin of safety" can narrow as a consequence (Figure 2.6); the net effect is that lifting with a flexed lumbar spine can concurrently lead to increased low-back (lumbar intervertebral joint) shear loading demands (Potvin et al. 1991a; Potvin et al. 1991b) and reduced low-back (intervertebral joint) loading capacity (Gallagher et al. 2005; Gunning et al. 2001; Howarth and Callaghan 2011). Performing the same lifting task (i.e., unchanged "external" task constraints), but with a neutral lumbar spine posture, can reduce the low-back shear loading demands and increase the low-back compressive loading capacity (McGill 1997).

2.6. Firefighter Low-Back Injury Prevention

It is unlikely that all firefighter low-back injuries can be prevented, but there are two types of low-back loading controls – *engineering* and *administrative* – that could conceivably reduce the number or severity of low-back overexertion injuries suffered during fireground operations. In alignment with the low-back injury framework adopted (McGill 1997) and the global objective of this thesis, effective controls would function to maintain a *margin of safety* at work by attenuating peak low-back loading demands on the fireground and/or increasing low-back loading capacity of firefighters. Reviewed below are some potential options available.

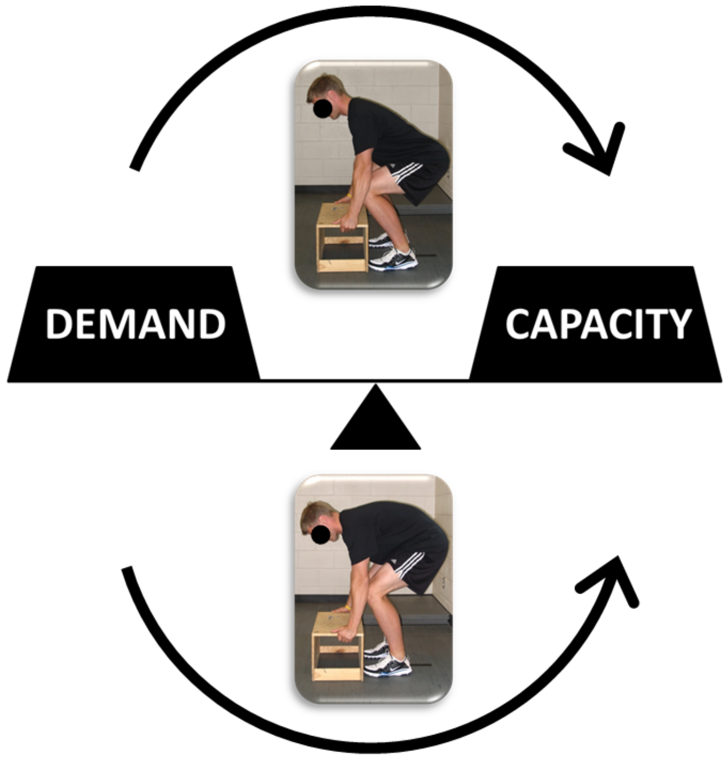

Figure 2.6. *Movement Matters!* Personal movement strategies can simultaneously impact low-back loading demands and capacity.

Engineering Controls

Engineering controls incorporate fundamental principles of biomechanics and ergonomics into the design or modification of work tasks and environments; they are typically *work-focused* in that they are conceived to *fit work to workers*. Engineering controls are preferred and prioritized for social and legal reasons and are most effectively and feasibly implemented in cases where the characteristics of work are predictable (e.g., manufacturing, office, and service jobs). Although task and environmental characteristics cannot always be controlled during fireground operations, assistive technologies and personal protective equipment can be ergonomically designed or modified to attenuate external physical demands at fire scenes (Coca et al. 2008; Coca et al. 2010; Coca et al. 2011; Griefahn et al. 2003; Hooper et al. 2001; Huang et al. 2009; Ilmarinen et al. 2008; Turner et al. 2010); exploiting such opportunities could reduce the potential for sustaining low-back injuries during fireground operations by preserving physical capacity in ways that permit objectives of non-modifiable tasks to be met without damaging low-back tissues. Specifically, results of the studies cited suggest that lighter and better-fitting personal protective equipment can reduce metabolic energy expenditure, perceived exertion, cardiorespiratory and thermal stress, and joint mobility restrictions, and could therefore delay or eliminate the need to employ potentially injurious adaptive movement strategies caused by fatigue, discomfort, or joint mobility restrictions. Moreover, low-back loading demands and capacity could also be altered in firefighters by reducing the total mass they must carry (i.e., reduced weight of tools, supplies, and protective clothing ensemble) and by modifying the distribution of this additional mass (e.g., alter size and shape of materials handled, design on-body load carriage systems, etc.) (Knapkik et al. 2004). Therefore, even though many fireground duties cannot be (re)designed, it is important to acknowledge the opportunities that exist to implement engineering controls into firefighter low-back injury prevention efforts.

Administrative Controls

In addition to the potential to implement engineering controls into occupational low-back injury prevention strategies for firefighters, administrative controls can also be utilized and may be particularly valuable in cases where specific fireground tasks cannot be modified. Administrative controls address the structure of work and the abilities of individuals who perform the work. Therefore, administrative controls can be either *work-focused* or *worker-focused*. Since firefighters perform irregular and often unplanned work as part of small teams comprised of individuals with specialized skills, it is often presumed impractical to design or modify the structure of their work through scheduling or job rotation schemes. However, it is important to consider how the overall structure of firefighters' work might contribute to low-back injury potential if attempting to design a comprehensive low-back injury program. One indication that work structure could affect fireground low-back injury potential is that fireground injury rates are over two times greater at night than they are during the day (Karter 2009). When firefighters respond to nocturnal alarms, their cognitive functioning and decision-making abilities can be acutely impaired (Elliot and Kuehl 2007), perhaps affecting their elected movement strategies (e.g., Sobeih et al. 2006) in ways that decrease their "margin of safety". Or, if obliged to perform tasks shortly after waking up, otherwise effective and non-injurious movement strategies could result in low-back injuries if intervertebral disc fluid absorption has reduced the low-back loading capacity (Gunning et al. 2001). Another indication that work structure could affect fireground low-back injury potential is that firefighters engage in exercise and training activities while on-duty. It is well-documented that firefighter injuries are commonly suffered during scheduled exercise- and training-related activities (Bylund and Björnstig 1999; Karter and Molis 2010; Loës and Jansson 2001; Poplin et al. 2011), but it is also important to consider that participation in these activities on-duty could alter low-back injury potential during subsequent emergency response operations (e.g., due to fatigue and/or reduced low-back loading capacity). Therefore, though it may be impractical to radically manipulate the work structure of firefighters, it may be worth considering in future research efforts some opportunities that do exist (e.g., modifying the types, frequencies, intensities, and durations of on-duty exercise and training activities).

Administrative controls to *fit workers to work* have long been employed in recruitment, hiring, and retention practices of firefighters, military service personnel, and law enforcement officers to ensure that members of the workforce attain and maintain the ability to safely and effectively perform their occupational duties (Sharkey and Davis 2008). Employee selection (e.g., pre-placement screening), job coaching and practice (e.g., technical and tactical training), and compulsory physical conditioning (e.g., exercise) are all examples *worker-focused* administrative controls available to reduce the potential for occupational low-back injuries in firefighters.

In North America, a collaborative effort between the International Association of Fire Chiefs (IAFC), International Association of Fire Fighters (IAFF), and ten North American fire departments and unions has resulted in the following three complimentary programs developed ultimately to enhance the health, safety, and well-being of firefighter recruits, incumbents, and retirees[1]:

1) Candidate Physical Ability Test (CPAT) – The CPAT program was developed and validated to assist in hiring firefighters, and it consists of a test and information related to recruitment, mentoring, and test administration and preparation. The test is comprised of eight simulated firefighting tasks, performed in a circuit-like fashion, which accurately and reliably represent critical tasks and essential job duties of firefighters. Passing the CPAT is accepted as a *bona fide* occupational requirement/qualification (Canada/United States), and thus test results can be used legally to make hiring decisions.
2) Joint Labour-Management Wellness-Fitness Initiative (WFI) – Through a partnership between the IAFC and IAFF, the WFI was conceived as a model that could be emulated to implement comprehensive wellness programs for active fire service personnel. Provided to participating departments through the WFI is a wealth of information pertaining to the following wellness program components: fitness evaluation; medical evaluation; injury prevention and rehabilitation; behavioural health; and data collection and reporting. Additional step-by-step strategies are recommended for successful implementation together with tools to assess the

[1] Descriptions are adapted from information available at the following websites (accessed on February 9, 2012):

http://www.iafc.org/wfi
http://www.iaff.org/hs/well

economic impact of implementation. Prevention of common firefighter (physical, mental, and emotional) health ailments in a main priority, but recommended systems and resources for the provision of comprehensive treatment and rehabilitation are also made available.

3) Peer Fitness Trainer (PFT) certification program – Developed in conjunction with the American Council on Exercise (ACE), the PFT certification program was designed to provide knowledge and practical skills needed for successful implementation of the WFI and CPAT. Interested personnel attend a five-day course during which PFT instructors educate attendees on basic principles and practices of behaviour modification, fitness testing and evaluation, and exercise prescription and progression. After completing the course, there is an option to challenge a certification examination. Educational resources (i.e., books, presentations, etc.) and testing materials (i.e., certification examination) were developed with the assistance of ACE content experts.

In developing the CPAT, WFI, and PFT programs, an explicit objective was to prevent the musculoskeletal injuries that compromise the work ability, health, safety, and well-being of firefighters themselves and the people they serve and protect. As reflected in the recommended WFI fitness testing and evaluation procedures, there is a particular emphasis on low-back injury prevention. Specifically, tests of low-back strength, flexibility, and endurance are recommended based on previous research linking poor performance on these tests with future low-back pain and injuries (Biering-Sørensen 1984; Cady et al. 1979; Cady et al. 1985).

More recent research has linked Functional Movement Screen™ (FMS) scores with a history of musculoskeletal complaints in fire and military service members (Keisel et al. 2009; Peate et al. 2007), and several participating fire departments now also include FMS tasks in WFI fitness testing and evaluation. As described in more detail in Chapter 5 and Appendix II, the FMS is comprised of seven tasks envisaged to identify general movement qualities that could constrain line-of-duty movement behaviour in hazardous ways. For example, though the FMS was not intended to be a "diagnostic" tool, poor performance on the Deep Squat test could indicate the presence of distal lower extremity joint dysfunction (e.g., ankle dorsiflexion restriction). If left unchecked, the number of available movement strategies to meet task objectives could be limited (e.g., inability to lift without bending or twisting the spine), and proximal areas of the body (e.g., intervertebral joints) could be rendered susceptible to overuse (e.g., disc herniation) or overexertion (e.g., posterior spinal ligament strain) injuries. Accordingly, if such qualities are detected in fitness evaluation and testing, individualized movement-focused exercise programs can be developed to enhance and maintain firefighter work ability and musculoskeletal durability. As described in more detail in Chapter 6, a *movement-centric* exercise approach differs from a more conventional *fitness-centric* approach in that priority is placed on grooving joint "load-sparing" patterns of coordination and control by attending to the *way* that exercises are performed (i.e., internal focus of attention). A more conventional *fitness-centric* exercise approach prioritizes improvements in strength, power, endurance, and flexibility by focusing on the *outcomes* or *effects* of their movements when exercising (i.e., external focus of attention). It could be argued that if both types of exercise yield similar improvements in physical fitness, but that a movement-centric approach also produces stable joint "load-sparing" movement adaptations, then a movement-centric approach would be a practical and effective worker-focused exercise-based low-back injury prevention strategy for firefighters. In Chapter 6, this notion was tested.

2.7. Summary

Though it is unlikely that all fireground low-back injuries can be prevented, continued efforts to reduce the number and impact of these injuries are warranted given the significant associated personal and societal costs. Opportunities still exist to implement engineering and administrative controls to attempt to *fit firefighting to firefighters*, but effective and practical worker-focused interventions designed to *fit firefighters to firefighting* will remain vital in future injury prevention programs. In adopting the occupational low-back injury framework presented by McGill (1997) and a *Movement Matters!* perspective (McGill 2009) (Chapter 3), a case was introduced for the notion that personal movement-impacting characteristics could constrain line-of-duty movement behaviours in ways that increase their low-back injury potential (Chapter 5). It was further introduced that if these characteristics could be identified using a whole-body movement screen (Chapter 5), movement-centric exercise approaches could be designed to enhance and maintain firefighter performance and durability (Chapter 6). These arguments motivated with research presented in the upcoming chapters.

CHAPTER 3

Low-Back Loading Demands during Simulated Firefighting Tasks – Inter-Subject Variation and the Impact of Fatigue and Gender

CHAPTER 3

Low-Back Loading during Simulated Firefighting Tasks – Inter-Subject Variation and the Impact of Fatigue and Gender

Summary

Background: Non-modifiable fireground duties are deemed hazardous for low-back health, but personal movement strategies could modulate low-back loading demands and injury potential. Objectives of this study were to quantify low-back loading demands during simulated firefighting tasks and to examine the impact of fatigue and gender on the peak loading response.

Methods: Ten men and 10 women performed a battery of laboratory-simulated firefighting tasks before and following repeated bouts of a fatiguing stair-climbing protocol. An EMG-assisted three-dimensional dynamic biomechanical model was used to compute peak L4/L5 joint forces during task performance.

Results: Peak low-back loading demands varied considerably between subjects and tasks, but 70% of all loading variables examined were of greater magnitudes in male subjects and 40% of all loading variables were of lower magnitudes in both males and females after stair-climbing. Some inter-subject variation in low-back loading was attributed to body size differences, but inter- and intra-individual variation in loading was also impacted by between- and within-subject differences in body movement and trunk muscle activation patterns.

Conclusions: Results of this study suggest that characteristics of individuals, tasks performed, and fatigue can influence peak low-back loading demands and injury potential in firefighters. Despite considerable inter-subject variation in the internal low-back loading response to fixed external task and environmental constraints, opportunities to attenuate low-back loading demands through movement behaviour modification alone may be limited to a subset of fireground activities.

3.1. Introduction

As indicated in Chapter 2, low-back injuries constitute a common (Karter and Molis 2010; Poplin et al. 2011) and very costly (NIST 2005) problem in the fire service. Despite the frequency and severity of low-back injuries sustained by firefighters, relatively few attempts have been made to quantify low-back loading demands during the performance of firefighting tasks. Since low-back injuries occur when the load-bearing capacity of the lumbar spine system is exceeded by imposed demands (McGill 1997), peak low-back load estimates provide injury prevention and rehabilitation researchers and practitioners with information that can be used directly in low-back injury risk analyses. Attenuating low-back loading demands through the implementation of engineering- or administrative-based ergonomic controls could reduce the potential for low-back injuries and pain reporting (Waters et al. 2006).

Although there is extensive literature describing the low-back loading patterns associated with manual material, tool, and patient handling tasks, results of such research cannot always or easily be extrapolated to predict the low-back loading demands imposed on firefighters. When firefighters perform comparable tasks on the fireground, for example, low-back compression and shear forces might differ from what would be expected because a self-contained breathing apparatus (SCBA) and turnout gear are worn. This on-body personal protective equipment could influence low-back loading demands by increasing total upper body mass and modifying its distribution, by altering trunk muscle activation patterns (Southard and Mirka 2007), and by causing individuals to adjust their whole-body movement strategies (Park et al. 2010) to maintain balance and posture (Punakallio et al. 2003; Sobeih et al. 2006) or to compensate for joint motion restrictions (Huck 1991). Previous efforts to quantify low-back loads during simulated firefighting tasks did not incorporate the potential effect of on-body load carriage (Lett and McGill 2006) and used biomechanical models that were insensitive to potential inter- and intra-individual variations in trunk muscle activation patterns on low-back load magnitude (Lusa et al. 1991; Lavender et al. 2000).

Also making it difficult to extrapolate previous research is that firefighters are often obliged to carry out essential job duties when physically fatigued. In a previous study comparing the performance of simulated firefighting tasks before and after a fatiguing stair-climbing protocol, Gregory et al. (2008) found that subjects exhibited increased lumbar spine flexion and decreased abdominal activation when fatigued. Although not examined in their study, it is likely that the internal low-back loading response was also impacted given inherent links between lumbar spine posture, trunk muscle activity, and low-back loading patterns (Reeves and Cholewicki 2003). Also, it has been demonstrated previously that occupational low-back loading demands vary in response to fatigue-related adaptations in movement strategies (Marras and Granata 1997; Dolan and Adams 1998; Dieën et al. 2001; Bonato et al. 2003). Since firefighter overexertion injuries are frequently sustained during fireground operations (Karter and Molis 2010), where some of their most physically demanding job duties are performed, it is important to account for the potential effect that fatigue could have on low-back loading demands.

Using a biomechanical modeling approach that was sensitive to inter- and intra-individual variations in movement and trunk muscle activation patterns, the primary objectives of this study were to quantify low-back loading demands during laboratory-simulated firefighting tasks and to examine the impact of fatigue on peak low-back compression and shear forces. Since gender differences in low-back loading have been observed when exposed to equivalent occupational task characteristics (Marras et al. 2002; Marras et al. 2003), a secondary objective of this study was to compare responses of men and women. It was hypothesized that peak low-back loading demands would differ between genders and that fatigue would alter the peak loading response.

3.2. Methods

Experimental Overview

Twenty volunteers performed a battery of laboratory-simulated firefighting tasks before and after repeated bouts of a fatiguing stair-climbing protocol (Figure 3.1). A three-dimensional dynamic EMG-assisted biomechanical model was used to quantify low-back loading demands during the task simulations. Peak L4/L5 compression and shear forces calculated prior to the first bout of stair-climbing were compared to those calculated following the last bout of stair-climbing that could be completed. Male subjects completed an average (standard deviation) of 2.4 (0.52) stair-climbing bouts, while females completed on average of 1.9 (0.57) bouts.

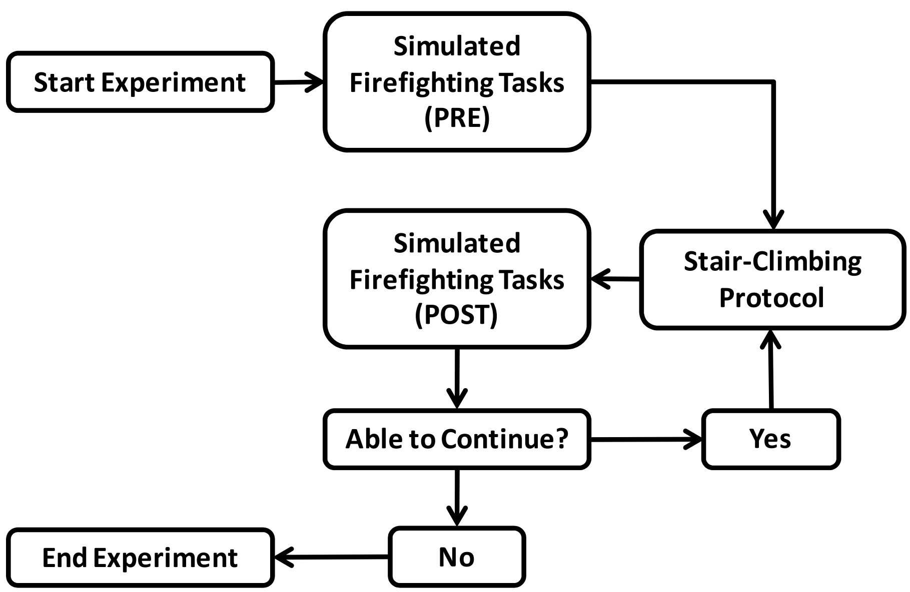

Figure 3.1. Laboratory-simulated firefighting tasks were performed before and after repeated bouts of a fatiguing stair-climbing protocol. Subjects were asked to perform the stair-climbing protocol as many times as they could. Peak L4/L5 joint forces calculated during the final set of firefighting tasks performed (POST) were compared to those calculated during the initial set (PRE).

Study Subjects

Ten men and 10 women took part in this study (Table 3.1). To ensure that subjects were able to meet the minimum demands of firefighting, only individuals who passed the Candidate Physical Ability Test (described below) within the previous six months were granted inclusion. Subjects signed informed consent documents that were approved by the University of Waterloo's Office of Research Ethics.

Table 3.1. Height, mass, and age of the 10 men and 10 women who participated.

	Subject	Height (m)	Mass (kg)	Age (yrs)
Men	M01	1.84	96.8	23
	M02	1.93	84.3	20
	M03	1.71	75.1	30
	M04	1.85	78.7	28
	M05	1.85	91.4	25
	M06	1.72	82.1	24
	M07	1.80	88.4	25
	M08	1.80	94.0	25
	M09	1.69	72.8	24
	M10	1.78	70.2	27
	Mean (SD)	1.80 (0.07)	83.4 (9.20)	25.1 (2.77)
Women	F01	1.71	62.1	23
	F02	1.72	62.2	19
	F03	1.64	67.6	30
	F04	1.72	72.5	24
	F05	1.66	62.7	27
	F06	1.70	75.4	34
	F07	1.72	89.8	21
	F08	1.63	78.1	20
	F09	1.72	71.6	23
	F10	1.65	60.1	27
	Mean (SD)	1.69 (0.04)	70.2 (9.28)	24.8 (4.71)

Candidate Physical Ability Test

A product of the Fire Service Joint Labor Management Wellness-Fitness Initiative, the Candidate Physical Ability Test (CPAT) was developed and validated through a cooperative effort between the International Association of Fire Chiefs, International Association of Fire Fighters, and 10 North American fire departments and unions to aid municipalities in the hiring of firefighters. The CPAT consists of eight simulated firefighting tasks – performed sequentially in a circuit-type fashion – which accurately and reliably represent the critical tasks and essential job duties of firefighters. Test elements include: stair-climb; hose advance; equipment carry; ladder raise and extension; forcible entry; search maze; victim rescue; and ceiling breach and pull. The entire CPAT is performed while wearing a 22.7 kg vest to simulate the mass of a SCBA and turnout gear; an additional 11.3 kg is attached to the vest during the stair-climbing task to represent the mass of a hose bundle. Individuals who can complete the entire circuit in less than the allotted time (10 minutes, 20 seconds) are considered physically able to perform critical tasks and essential job duties at a fire scene. In North America, passing the CPAT is accepted as a *bona fide* occupational requirement (Canada) or qualification (United States).

Given that CPAT tasks are considered representative of critical and essential duties performed on the fireground, laboratory tasks were designed to represent components of CPAT tasks hypothesized to impose the greatest low-back mechanical demands on performers. Task selection was primarily influenced by the postures assumed and equipment handled during the CPAT. However, constraints imposed by the instrumentation used (e.g., force platforms, optoelectronic motion capture system, etc.), space available, and props required (e.g., sledgehammer, pike pole, etc.) were also considered when designing laboratory tasks.

Laboratory-Simulated Firefighting Tasks

After reviewing CPAT training and instructional materials (official manuals and videos) and observing live practice and testing sessions conducted over a two-week period at a licensed testing facility (University of Waterloo Fitness Unit), the following components of the CPAT were mocked-up in a biomechanics laboratory:

Hose Advance. Through a system of pulleys and cables, a rope was attached to a 30 kg mass. Measurements made at the CPAT testing facility were used to select the mass; the mass was selected such that the force acting on the body during the initiation of the laboratory-based hose advance task approximated the same quantity measured at the CPAT testing facility using a force transducer (Chatillon Ergonomic Gauge, Ametek, Inc., Berwyn, PA, United States). As instructed during the CPAT, subjects were asked to place the rope over their preferred shoulder and walk across the force platforms at a self-selected walking speed (Figure 3.2). The initiation and single-support phase (right foot) of a stride was captured for analyses.

Figure 3.2. Hose advance. Both the initiation and a single-support phase (right foot) were captured for analyses.

Kneeling Hose Pull. Also through a pulley-cable system, subjects executed a hand-over-hand rope pull while half-kneeling (Figure 3.3). All subjects performed the task while kneeling on their left side. Resistance (30 kg) was selected to approximate the equivalent quantity measured at the CPAT testing facility.

Figure 3.3. Kneeling hand-over-hand hose pull. All participants performed this task while kneeling on their left side.

Equipment Lift and Carry. Two components of the CPAT equipment carry task were simulated in the laboratory. First, subjects were asked to lift a 12.7 kg mass with their right hand from the floor beside them (Figure 3.4a). Second, subjects were asked to carry 12.7 kg in each hand while walking across the force platforms; data were captured during the single-support phase (right foot) of the stride (Figure 3.4b).

Figure 3.4. a) Equipment lift: 12.7 kg mass was lifted with the right hand from the floor; b) Equipment carry.

Forcible Entry. Subjects used a 4.5 kg sledgehammer to strike five times (in succession) a 45.5 kg sand-filled heavy bag that was hanging from the laboratory ceiling (Figure 3.5). Subjects were instructed to strike the bag as though they were trying to enter through a lodged door as quickly and efficiently as possible whilst maintaining body control, striking accuracy, and precision. To prohibit the heavy bag from swinging, a research assistant grasped handles affixed to the heavy bag.

Victim Rescue. Because it was not possible to drag a mannequin in the laboratory without interfering with force platform measurements, subjects were asked instead to pull on a handle that was fastened to a pulley-cable system. To simulate the initiation of the CPAT victim rescue task, the cable was routed under a bar so that a force was applied to the hands of subjects in a way that mimicked that measured at the commencement of the original CPAT task (Figure 3.6). The “dragging” component of the CPAT victim rescue task was also simulated by asking subjects to pull on the same handle while stepping backward across the force platforms (Figure 3.7); however, the bar was removed for this dragging component. Again, the mass pulled was selected to match the equivalent force measured at the CPAT testing facility.

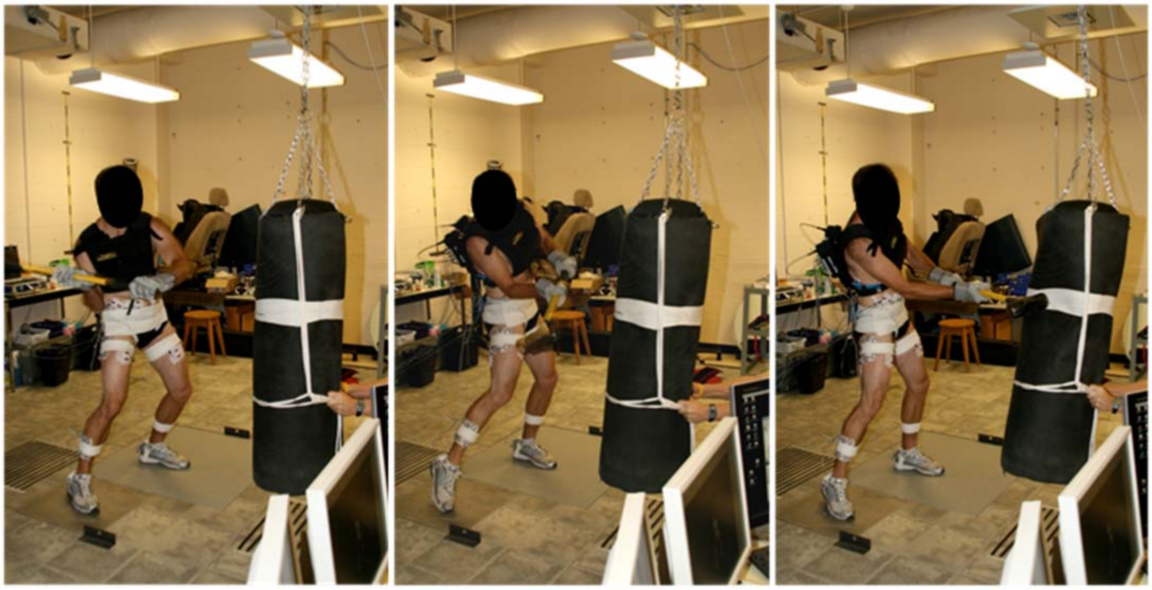

Figure 3.5. Forcible entry task. A 4.5 kg sledgehammer was used to strike a 45.5 kg heavy bag five times.

Figure 3.6. Victim rescue task. Pictured here is the initiation component of the task.

Figure 3.7. Victim rescue task. Pictured here is the "dragging" component of the task, wherein subjects pulled on a cable system while backing-up across the force platforms.

Ceiling Breach and Pull. Subjects simulated ceiling breaching and pulling tasks while handling a dowel that was attached to a system of pulleys and cabling. The breach component of the CPAT task was simulated by moving the dowel in an upward direction with its downward-facing end fastened to a cable below (Figure 3.8); a bar was used to guide the cable in the desired direction. The ceiling pull component of the CPAT task was simulated by pulling down on the dowel while its upward-facing end was fastened to a cable overhead (Figure 3.8). Subjects were instructed to simulate, to the best of their recollection, the movements performed during the CPAT ceiling breach and pull tasks. Masses were selected to approximate force measurements made at the CPAT testing facility.

While performing the laboratory tasks, subjects wore gloves and a 22.7 kg vest to simulate the mass of turnout gear and a SCBA. Consistent with how the CPAT is conducted, laboratory tasks were performed in a circuit-type fashion and in the same order by all subjects. A minimum of three successful trials of each task were recorded to increase the stability of the measurements.

Figure 3.8. Ceiling breach and pull.

Stair-Climbing Protocol

The stair-climbing protocol was conducted as described in the official CPAT Administration and Procedures Guide. With an additional 11.3 kg mass added to the 22.7 kg vest (total vest mass = 34 kg), subjects climbed on a stair-climbing machine for 3 minutes at a rate of 60 steps/minute (StairMaster StepMill 7000PT, Nautilus Inc., Vancouver, WA, USA) (Figure 3.9). Subjects were not permitted to use the hand-rails.

Instrumentation

Pairs of pre-gelled self-adhesive surface EMG recording electrodes (Medi-Trace, Kendall-LTP, Chicopee, MA, United States) were applied to the skin over the following seven bilateral trunk muscle groups: upper erector spinae (≈ 5 cm lateral to the T9 spinous process); lower erector spinae (≈ 3 cm lateral to the L3 spinous process); rectus abdominis (≈ 3 cm lateral to umbilicus); external abdominal obliques (≈ 15 cm lateral to umbilicus); internal abdominal obliques (midway between the anterior superior iliac spine and symphysis pubis); and latissimus dorsi (lateral to the T9 spinous process over the muscle belly) (Cholewicki and McGill 1996). EMG signals were band-pass filtered (10-1000 Hz) and differentially amplified (CMRR = 115 dB at 60 Hz; input impedance = 10 GΩ) (AMT-16, Bortec Biomedical Ltd., Calgary, AB, Canada) prior to analog-to-digital conversion at a rate of 2048 Hz using an Optotrak® Data Acquisition Unit (ODAU II, Northern Digital Inc., Waterloo, ON, Canada).

Clusters of five or six Optotrak® Smart Markers (Northern Digital Inc., Waterloo, ON, Canada) affixed to custom-molded rigid bodies were secured to the feet, shanks, thighs, pelvis, and thorax of subjects using double-sided tape and Velcro® straps. Marker position data were collected at a rate of 32 Hz using a six-sensor Optotrak Certus® motion capture system (Northern Digital Inc., Waterloo, ON, Canada).

Signals from two ground-mounted AMTI force platforms (Advanced Mechanical Technology, Inc., Watertown, MA, United States) were collected at a rate of 2048 Hz.

EMG, force platform, and marker data were temporally synchronized using an Optotrak® Data Acquisition Unit (ODAU II, Northern Digital Inc., Waterloo, ON, Canada).

Data Processing and Analyses

Before performing laboratory tasks, EMG signals were collected while subjects performed maximum voluntary isometric contractions (MVICs) of the monitored muscle groups. Briefly, with their lower body restrained, subjects performed a series of standardized maximal trunk flexion, extension, lateral bend, and axial twist exertions while research assistants provided a matching resistance (McGill 1991) (Figure 3.10). Two or three repetitions of each exertion were performed; one-minute of rest was provided between consecutive exertions.

Figure 3.9. Stair-climbing protocol. Subjects climbed for 3 minutes at a rate of 60 steps/minute.

Figure 3.10. EMG normalization tasks for the trunk extensor muscles (top panel) and abdominal wall muscles (bottom panel).

EMG signals were full wave rectified and digitally low-pass filtered (Butterworth, second-order, single pass) with a 2.5 Hz cut-off frequency to produce a linear envelope (Brereton and McGill 1998). EMG signals collected during the experimental trials were normalized to the peak amplitudes detected during the MVIC trials.

An orthogonal coordinate system was derived based on each rigid body marker cluster using commercial software (NDI 6D Architect™, Northern Digital Inc., Waterloo, ON, Canada) so that positions of anatomically meaningful medial and lateral segment endpoints could be tracked throughout the experimental trials. The location of each segment endpoint was described with respect to the rigid body coordinate system by digitizing the appropriate landmarks during a quiet standing trial (Figure 3.11). Anatomical landmark data were then used in Visual3D™ software (Version 4, C-Motion, Inc., Germantown, MD, United States) to create body segment-fixed coordinate systems. Proximal endpoints of the thighs and shanks were located "functionally" by having subjects perform 8 to 10 repetitions of controlled "open" kinematic chain rotations about the hips and knees (Begon et al. 2007). Calculations of these functionally-derived segment endpoints were also performed in Visual3D™ (Schwartz and Rozumalski 2005) as was the creation of landmarks used to define the frontal plane of the shank and thigh segments. Effectively, the procedures above yielded a linked-segment model of the body.

Positions and orientations of each modeled body segment were calculated using the default six degree-of-freedom optimal tracking algorithms in Visual3D™ (c.f., Cereatti et al. 2006). Angular displacements of modeled "joints" (e.g., knee) were represented as the orientation of distal segments (e.g., shank) with respect to their adjacent proximal segments (e.g., thigh) (Woltring 1991); the rotation matrix describing the relative three-dimensional orientation between adjoining segments was decomposed using the following sequence of rotations (Cole et al. 1993): flexion/extension (sagittal plane) $\rightarrow$ abduction/adduction (frontal plane) $\rightarrow$ axial rotation (transverse plane). Angular displacement data were padded using an end-point reflection method (Smith 1989) and then low-pass filtered using a zero-lag, fourth-order digital Butterworth filter. Residual analyses (Winter 2005) were used in selecting a filter cut-off frequency of 3 Hz.

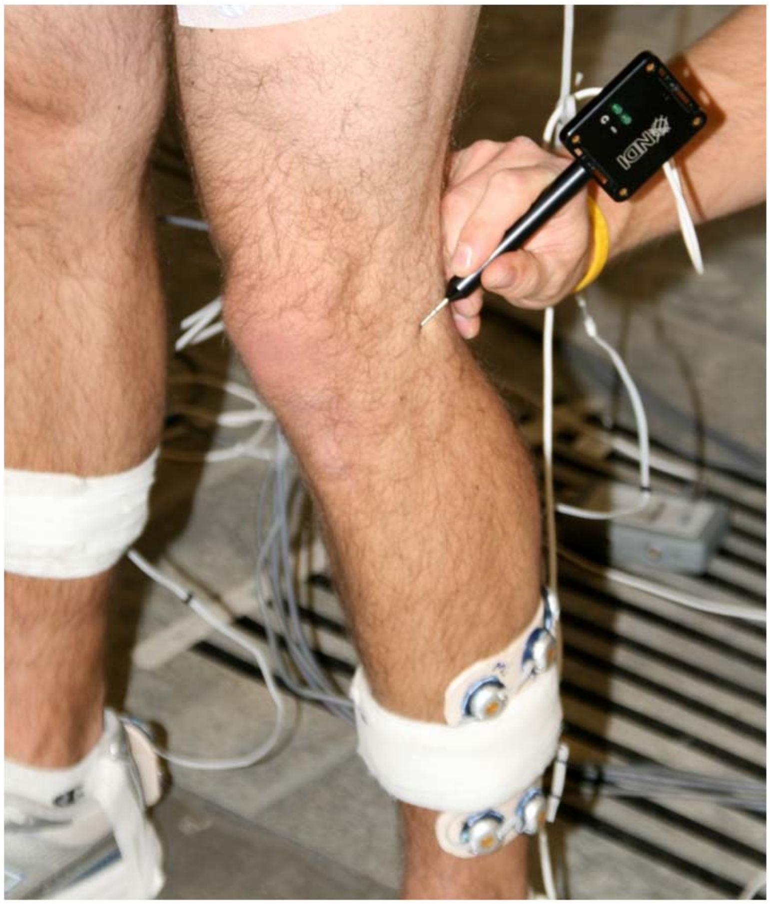

Figure 3.11. Anatomical landmark digitization procedure. The tip of the digitizing pointer was placed over a bony landmark (lateral tibial condyle) and its position was calculated with respect to the local coordinate system of the marker cluster on the shank.

Using a "bottom up" inverse dynamics approach (c.f., Faber et al. 2010), Visual3D™ was also used to compute instantaneous reaction forces and net joint moments of force about the origin of the pelvis segment coordinate system during task performance. The origin of the pelvis segment coordinate system was calculated as the mid-point between iliac crests, and a least-squares plane fit to the locations of iliac crests and greater trochanters represented the frontal plane of the pelvis segment. The component of the reaction force acting normal to the frontal plane of the pelvis was assumed equivalent to anterior/posterior reaction shear force acting at the L4/L5 joint, and frontal plane component of the net joint moment was assumed equivalent to the lateral bend moment about the L4/L5 joint. Body segment mass and inertial parameters incorporated in the inverse dynamics analyses were calculated based on the default procedures in Visual3D™ (i.e., using published anthropometric data (Dempster 1955) and geometric models of body segments). Relevant quantities derived from the inverse dynamical linked-segment model analyses (i.e., net L4/L5 joint moments, L4/L5 reaction forces, and lumbar spine angles) were then incorporated into an EMG-assisted musculoskeletal model of the lumbar torso to quantify L4/L5 joint ("bone-on-bone") compression and shear forces (Cholewicki and McGill 1996).

Although the musculoskeletal modeling process employed is well-documented in the literature (Cholewicki and McGill 1996; McGill 1992; McGill and Norman 1986), a description of the approach is included here to aid in the interpretation of results. Contributions of passive tissues (e.g., ligaments, intervertebral discs, and gut) to the net L4/L5 joint moment were estimated first based on the angular displacement of the ribcage with respect to the pelvis (i.e., lumbar spine angle). By assuming that each intervertebral joint contributes a constant proportion of the total lumbar spine angular displacement (McGill and Norman 1986; McGill 1992), the lumped passive tissue moment contributions were calculated based on joint displacement-load relationships (McGill et al. 1994). The remaining moment (difference between the net L4/L5 joint moment and the estimated moment contributed by passive tissues about the L4/L5 joint) was then partitioned amongst the muscles by combining the anatomical model of Cholewicki and McGill (1996) with distribution-moment (DM) equations (Ma and Zahalak 1991) to compute muscle forces and stiffness' (Cholewicki and McGill 1995). Normalized linear enveloped EMG data were then used in the DM equations to represent muscle activation; muscle groups not directly accessible for surface EMG recording were assigned activation profiles from anatomically and functionally similar muscle groups (McGill et al. 1996a). Muscle attachment data reported by Cholewicki and McGill (1996) were combined with measurements of three-dimensional lumbar spine kinematics to calculate instantaneous muscle lengths and contraction velocities used in the DM equations. Each muscle's maximum force-producing capability (i.e., maximum muscle stress multiplied by the physiological cross-sectional area) was incorporated in the DM equations also based on data reported by Cholewicki and McGill (1996).

To compensate for potential errors associated with the transformation of EMG signals into estimates of muscle force and stiffness (based on DM equations), a calibration procedure was used to "fit" the model to each subject. Specifically, a single gain factor *G* was computed for each subject by finding the value for *G* that minimized the total sum of squared differences between instantaneous estimates of the net L4/L5 joint moment derived from the linked-segment model ($\mathbf{M}_{LSM}$) and the equivalent quantity predicted by the EMG-driven lumbar torso model ($\mathbf{M}_{EMG}$) (Equation 3.1). Data collected over the duration of each trial collected (f = 1 to F frames) and each of three anatomical axes (a = 1 to A representing the flexion/extension [a = 1], lateral bend [a = 2], and axial twist [a = 3] axes) were used to find *G*. By multiplying instantaneous muscle force and stiffness estimates (derived initially from DM equations) by *G*, muscle force and stiffness values were adjusted at each instant in time during the experimental trials to account for between-subject differences in factors that can influence EMG-to-force transformations (e.g., muscle morphology) (Cholewicki et al. 1995). The gain factor was "common" in that the same *G* was used to scale all muscle force estimates within an individual, but *G* varied between individuals, tasks, and over time to ensure that the magnitudes of $\mathbf{M}_{\mathrm{LSM}}$ and $\mathbf{M}_{\mathrm{EMG}}$ were "matched" as closely as possible throughout the experiment.

$$\sum_{f=1,a=1}^{F,A}\left(\mathrm{M}_{LSM(a,f)} - G\mathrm{M}_{EMG(a,f)}\right)^2 = \min \qquad \text{(Equation 3.1)}$$

Passive tissue and adjusted muscle forces acting at the L4/L5 joint were added to the LSM-derived L4/L5 reaction forces to quantify L4/L5 joint compression and shear forces during the experimental tasks (Potvin et al. 1991a; Potvin et al. 1991b). Hence, L4/L5 joint forces incorporated contributions from passive tissue forces, muscle forces, external forces and moments (acting at the interface between the force platforms and feet), and the gravitational-inertial forces associated with moving body segments.

From all kinematic, kinetic, and EMG waveforms, global maximum and minimum values were extracted and the “peak” was defined as the greater (absolute value) of the two. Lumbar spine motion was defined as the difference between the maximum and minimum angle. Means for each measure (averaged across 3 trials) constituted the dependent variables in statistical analyses.

Statistical Analyses

Using the general linear model procedure in SAS system software (Windows Version 9.1.3 with Service Pack 4, SAS Institute Inc., Cary, NC, United States), dependent variables were compared between men and women and over time. Data from each task were analyzed independently. Least-square means were computed if gender×time effects were statistically significant; adjustments for multiple comparisons were performed using the Tukey method. p-values less than 0.05 were considered to be statistically significant; however, cases where weak evidence against the null hypothesis was offered (i.e., $p < 0.1$) were considered noteworthy.

3.3. Results

Global Findings

Attributable to inter-individual differences in body sizes, trunk muscle activation levels, and movement patterns was the finding that peak L4/L5 joint forces varied considerably between subjects (Table 3.2). To help interpret the implications of this inter-individual variation, the number of cases where a subject exceeded recommended exposure limits is summarized in Table 3.3.

Table 3.2. Minimum and maximum peak L4/L5 joint compression, A/P shear, and M/L forces (N) calculated during simulated firefighting tasks. Data from all men (N = 10) and women (N = 10) are included.

Task	Time	Compression		A/P Shear		M/L Shear	
		Min (N)	Max (N)	Min (N)	Max (N)	Min (N)	Max (N)
Hose Advance (initiation)	Pre	1473	4788	150	1017	-626	316
	Post	1729	4800	-200	978	-375	190
Hose Advance (stride)	Pre	1570	4307	102	1053	-454	188
	Post	1092	4816	-135	1375	-370	86
Kneeling Hose Pull	Pre	2087	4717	277	1713	-667	692
	Post	1891	4994	-231	1491	-617	824
Equipment Lift	Pre	3732	7910	524	2515	-209	423
	Post	3018	6965	323	2184	-223	421
Equipment Carry	Pre	1462	3688	-255	557	-217	226
	Post	1397	4405	-279	698	-169	197
Forcible Entry	Pre	3841	14911	436	2675	-509	1263
	Post	3453	12390	569	1906	-415	1018
Victim Rescue (drag)	Pre	2578	5287	241	1651	-215	376
	Post	2312	5368	-162	1778	-235	309
Victim Rescue (initiation)	Pre	3276	8104	-255	2505	-192	198
	Post	2680	7321	-434	2066	-171	237
Ceiling Breach	Pre	1945	7489	123	1451	-380	758
	Post	2765	6932	-424	1493	-362	652
Ceiling Pull	Pre	1963	5064	-402	1300	-406	371
	Post	1494	4448	-431	1027	-371	636

Table 3.3. Total number of subjects (number of females) who experienced L4/L5 joint to compression, A/P shear, and M/L shear forces in excess of recommended exposure limits.

Task	Time	Compression[1]		A/P Shear[2]		M/L Shear[3]	
		AL†	MPL‡	AL†	MPL‡	AL†	MPL‡
		(> 3.4 kN)	(> 6.4 kN)	(> 0.5 kN)	(> 1.0 kN)	(> 0.5 kN)	(> 1.0 kN)
Hose Advance (initiation)	Pre	5 (3)	—	7 (3)	1 (0)	1 (0)	—
	Post	3 (2)	—	4 (2)	—	—	—
Hose Advance (stride)	Pre	3 (1)	—	3 (1)	1 (0)	—	—
	Post	3 (0)	—	3 (0)	1 (0)	—	—
Kneeling Hose Pull	Pre	10 (4)	—	16 (6)	6 (2)	5 (2)	—
	Post	7 (1)	—	16 (7)	3 (1)	4 (1)	—
Equipment Lift	Pre	20 (10)	1 (0)	20 (10)	14 (5)	—	—
	Post	19 (9)	2 (0)	19 (9)	13 (4)	—	—
Equipment Carry	Pre	1 (0)	—	1 (0)	—	—	—
	Post	1 (0)	—	2 (0)	—	—	—
Forcible Entry	Pre	20 (10)	6 (2)	18 (8)	9 (2)	13 (5)	1 (0)
	Post	20 (10)	4 (0)	20 (10)	6 (0)	9 (3)	1 (0)
Victim Rescue (drag)	Pre	11 (4)	—	11 (3)	5 (1)	—	—
	Post	11 (3)	—	9 (1)	2 (0)	—	—
Victim Rescue (initiation)	Pre	19 (9)	3 (0)	16 (6)	9 (3)	—	—
	Post	19 (9)	2 (0)	15 (5)	10 (3)	—	—
Ceiling Breach	Pre	14 (7)	1 (0)	12 (5)	4 (2)	1 (0)	—
	Post	16 (6)	1 (0)	14 (5)	4 (2)	1 (0)	—
Ceiling Pull	Pre	11 (3)	—	9 (3)	1 (0)	—	—
	Post	9 (1)	—	7 (1)	2 (0)	1 (0)	—

†Spinal load magnitudes in excess of the Action Limit (AL) are considered potentially hazardous for some workers.

‡ Spinal load magnitudes in excess of the Maximum Permissible Limit (MPL) are considered hazardous for most workers.

[1]Limits based on recommendations made by the National Institute of Occupational Safety and Health (NIOSH 1981) and Waters et al. (1993).

[2]Limits based on recommendations made by McGill et al. (1998).

[3]There are no published AL or MPL values for M/L shear forces. For the purpose of this exercise and consistent with other opinions (Marras 2008), it was assumed that limits for A/P shear would suffice.

Despite the fact that low-back load magnitudes varied considerably between subjects, peak L4/L5 joint compression, A/P shear, and M/L shear forces in males were of significantly greater magnitudes in 9, 6, and 5 of the 10 tasks, respectively. When peak low-back load magnitudes were normalized with respect to subject bodyweight, statistically significant gender-based differences in peak L4/L5 joint compression, A/P shear, and M/L shear force magnitudes were then detected in only 3, 1, and 3 of the tasks, respectively (Tables 3.4 to 3.6). It should be noted, however, that if a *p*-value of less than 0.1 were accepted as a cut-point (instead of $p < 0.05$), gender differences in bodyweight normalized peak L4/L5 joint compression, A/P shear, and M/L shear forces would have remained in 4, 6, and 4 of the tasks, respectively (Tables 3.4 to 3.6).

In 6, 3, and 2 of the 10 simulated firefighting tasks, peak L4/L5 compression, A/P shear, and M/L shear forces (respectively) were of significantly lower magnitudes when performed after the final stair-climbing protocol (POST trials) than those achieved when performing tasks before the first bout of stair-climbing (PRE trials).

When investigating why peak L4/L5 forces typically differed between genders and over time, it was found that common gain factor values (*G* in the Methods section) could be used to explain the low-back loading responses observed. Gain factor magnitudes were significantly greater for males than for females, and POST trial gain factors were of significantly lower magnitudes than the PRE trial gains in 6 out of 10 tasks (Table 3.7). Because trunk muscles contribute significantly to the total (“bone-on-bone”) lumbar intervertebral joint force magnitude (Potvin et al. 1991a), peak L4/L5 forces are highly sensitive to the amount of “adjustment” made to muscle force calculations through the application of a gain factor. From a computational standpoint, it was thus easy to understand why a number of gender- and time-based differences were observed. A physiological explanation for the differences is less clear based solely on gain factor values, but an examination of the task-by-task findings below provides some insight into the underlying mechanisms. (EMG data referred to in the following sections are summarized in Appendix I.)

Table 3.4. Mean (SEM) peak L4/L5 joint compression forces expressed as a percentage of subject bodyweight (BW). Data from all male (N = 10) and female (N = 10) subjects are included.

Task	Time	Compression (% BW)				p-values†		
		Males		Females		gender	time	gender×time
Hose Advance (initiation)	Pre	346	(26.5)	410	(42.2)	0.2067	0.0262	0.6561
	Post	311	(30.5)	359	(30.7)			
Hose Advance (stride)	Pre	377	(24.2)	324	(21.8)	0.0195	0.0023	0.0526
	Post	358	(36.4)	242	(13.7)			
Kneeling Hose Pull	Pre	444	(19.5)	438	(30.1)	0.4666	0.5347	0.1846
	Post	453	(20.9)	414	(20.6)			
Equipment Lift	Pre	693	(28.2)	651	(21.5)	0.0697	0.1943	0.0911
	Post	699	(28.0)	609	(26.0)			
Equipment Carry	Pre	316	(15.2)	298	(18.5)	0.1314	0.0038	0.0518
	Post	306	(25.9)	246	(11.2)			
Forcible Entry	Pre	904	(110)	798	(78.9)	0.2506	0.1056	0.4369
	Post	837	(89.9)	674	(38.1)			
Victim Rescue (drag)	Pre	515	(33.6)	496	(31.4)	0.3063	0.0060	0.2222
	Post	486	(28.5)	427	(19.3)			
Victim Rescue (initiation)	Pre	762	(29.9)	652	(29.6)	0.0630	0.1173	0.3209
	Post	719	(30.5)	642	(47.6)			
Ceiling Breach	Pre	594	(58.2)	552	(34.7)	0.3960	0.2597	0.7031
	Post	640	(48.1)	575	(50.3)			
Ceiling Pull	Pre	507	(35.4)	397	(20.7)	0.0170	0.0148	0.8426
	Post	471	(37.3)	366	(23.9)			

†A general linear model ANOVA with one between-subject factor (gender: Males vs. Females) and one within-subject factor (time: Pre vs. Post) was performed to compare low-back load magnitudes.

Table 3.5. Mean (SEM) peak L4/L5 joint anterior/posterior shear forces expressed as a percentage of subject bodyweight (BW). Data from all male (N = 10) and female (N = 10) subjects are included.

Task	Time	A/P Shear (% BW)				p-values†		
		Males		Females		gender	time	gender×time
Hose Advance (initiation)	Pre	55	(9.4)	65	(10.5)	0.4352	0.0552	0.9469
	Post	45	(6.8)	55	(10.9)			
Hose Advance (stride)	Pre	58	(9.2)	41	(7.9)	0.0899	0.3376	0.3102
	Post	58	(14.9)	30	(4.5)			
Kneeling Hose Pull	Pre	121	(9.1)	95	(14.2)	0.2092	0.1792	0.3970
	Post	105	(11.6)	92	(11.8)			
Equipment Lift	Pre	222	(18.6)	141	(12.7)	0.0020	0.0084	0.3733
	Post	190	(15.6)	125	(14.1)			
Equipment Carry	Pre	31	(5.5)	26	(3.2)	0.4072	0.2467	0.7969
	Post	35	(8.3)	28	(2.7)			
Forcible Entry	Pre	157	(19.9)	128	(19.1)	0.0825	0.1196	0.6653
	Post	144	(12.9)	104	(5.2)			
Victim Rescue (drag)	Pre	110	(16.2)	81	(13.2)	0.0639	0.0133	0.4660
	Post	95	(15.9)	54	(7.5)			
Victim Rescue (initiation)	Pre	154	(17.4)	100	(17.0)	0.0623	0.5342	0.5541
	Post	154	(14.7)	111	(23.3)			
Ceiling Breach	Pre	85	(13.8)	83	(16.0)	0.9182	0.1669	0.9650
	Post	96	(11.5)	93	(23.8)			
Ceiling Pull	Pre	85	(13.0)	56	(8.9)	0.0590	0.0644	0.6242
	Post	74	(11.4)	49	(5.0)			

†A general linear model ANOVA with one between-subject factor (gender: Males vs. Females) and one within-subject factor (time: Pre vs. Post) was performed to compare low-back load magnitudes.

Table 3.6. Mean (SEM) peak L4/L5 joint medial/lateral shear forces expressed as a percentage of subject bodyweight (BW). Data from all male (N = 10) and female (N = 10) subjects are included.

Task	Time	M/L Shear (% BW)				p-values†		
		Males		Females		gender	time	gender×time
Hose Advance (initiation)	Pre	36	(5.4)	34	(4.8)	0.8446	0.0070	0.6169
	Post	26	(2.8)	26	(2.9)			
Hose Advance (stride)	Pre	27	(4.2)	16	(2.7)	0.0072	0.0422	0.8450
	Post	22	(3.7)	10	(1.3)			
Kneeling Hose Pull	Pre	51	(5.1)	51	(5.4)	0.6502	0.7426	0.5079
	Post	49	(6.7)	55	(5.7)			
Equipment Lift	Pre	34	(4.1)	31	(3.4)	0.5623	0.4111	0.8544
	Post	32	(3.7)	29	(3.7)			
Equipment Carry	Pre	14	(2.1)	15	(2.9)	0.5721	0.6730	0.9006
	Post	13	(1.2)	15	(2.0)			
Forcible Entry	Pre	91	(9.7)	66	(7.6)	0.0722	0.2775	0.5257
	Post	83	(9.8)	64	(7.3)			
Victim Rescue (drag)	Pre	19	(1.4)	34	(3.8)	0.0052	0.5085	0.1677
	Post	23	(2.7)	33	(3.5)			
Victim Rescue (initiation)	Pre	19	(1.6)	20	(1.0)	0.9774	0.6831	0.4224
	Post	20	(1.2)	19	(1.5)			
Ceiling Breach	Pre	31	(6.2)	31	(4.9)	0.5417	0.5118	0.1233
	Post	33	(4.6)	25	(3.2)			
Ceiling Pull	Pre	32	(4.3)	24	(1.3)	0.0463	0.4356	0.7931
	Post	34	(4.8)	28	(2.5)			

†A general linear model ANOVA with one between-subject factor (gender: Males vs. Females) and one within-subject factor (time: Pre vs. Post) was performed to compare low-back load magnitudes.

Table 3.7. Common gain factor values derived to calibrate the EMG-driven musculoskeletal lumbar torso model. Mean (standard error of the mean) values are presented.

Task	Time	Common Gain Factor				*p*-value†		
		Males		Females		gender	time	gender×time
Hose Advance (initiation)	Pre	1.29	(0.14)	0.86	(0.09)	0.0123	0.0032	0.3643
	Post	1.00	(0.10)	0.69	(0.08)			
Hose Advance (stride)	Pre	1.40	(0.22)	0.75	(0.05)	0.0022	0.0014	0.8809
	Post	1.19	(0.16)	0.51	(0.04)			
Kneeling Hose Pull	Pre	1.16	(0.10)	0.79	(0.05)	0.0020	0.0026	0.8927
	Post	1.06	(0.10)	0.68	(0.04)			
Equipment Lift	Pre	1.60	(0.09)	1.01	(0.04)	<0.0001	0.0056	0.1519
	Post	1.54	(0.09)	0.84	(0.05)			
Equipment Carry	Pre	1.23	(0.22)	0.75	(0.07)	0.0339	<0.0001	0.2534
	Post	1.06	(0.23)	0.49	(0.04)			
Forcible Entry	Pre	1.24	(0.09)	0.82	(0.04)	0.0002	0.0578	0.3965
	Post	1.09	(0.06)	0.71	(0.04)			
Victim Rescue (drag)	Pre	1.14	(0.12)	0.71	(0.05)	0.0020	0.0071	0.6970
	Post	1.03	(0.10)	0.63	(0.04)			
Victim Rescue (initiation)	Pre	1.40	(0.09)	0.88	(0.04)	<0.0001	0.6421	0.0723
	Post	1.44	(0.09)	0.81	(0.06)			
Ceiling Breach	Pre	1.28	(0.17)	0.85	(0.05)	0.0078	0.9754	0.3842
	Post	1.33	(0.16)	0.79	(0.05)			
Ceiling Pull	Pre	1.08	(0.04)	0.63	(0.05)	<0.0001	0.9788	0.3578
	Post	1.09	(0.04)	0.61	(0.05)			

†A general linear model ANOVA with one between-subject factor (gender: Males vs. Females) and one within-subject factor (time: Pre vs. Post) was performed to compare gain factor magnitudes.

Hose Advance (initiation)

In comparison to males, females tended to display greater activation levels ($p \leq 0.0512$) in abdominal oblique muscles (REO, RIO, LIO, LEO). However, there were no gender-based differences detected in peak L4/L5 compression ($p = 0.9480$), A/P shear ($p = 0.8830$), and M/L shear ($p = 0.2631$) forces (Figure 3.12). When coupled with the finding that no gender-based differences were found in L4/L5 net moment magnitudes ($p \geq 0.1122$), the observation that females used greater levels of muscle activation (than did males) to produce equivalent L4/L5 moments largely explained the gender-based differences in gain factor magnitudes for this task (Table 3.7). This was not unexpected given that gender-based differences in the moment-producing capabilities of trunk muscles (e.g., McGill 1991) have been linked with male-female differences in physiological cross-sectional areas and lines-of-action (Jorgensen et al. 2001; Marras et al. 2001b). However, at least part of the gender-based differences in gain factor magnitudes could also have been related to the observation that male subjects exhibited 6° more lumbar twist motion ($p = 0.0402$) than did females when executing initiation component of the Hose Advance task. It could be argued that a difference of this magnitude was not likely a major contributor to gender-based differences in spine loading responses, but it is possible that this difference contributed additional between-gender discrepancies in muscle force- and moment-producing capabilities due to lumbar motion-induced changes in muscle lengths, velocities, or lines-of-action.

Following the stair-climbing protocol, peak L4/L5 joint forces were 11.6% (330 N), 16.5% (74 N), and 26.3% (69 N) lower in the compressive ($p = 0.0233$), A/P shear ($p = 0.0457$), and M/L shear ($p = 0.0076$) directions, respectively (Figure 3.12). Again, these changes were related to the fact that POST trial gain factors were significantly lower than PRE trial gain factors (Table 3.7). Although somewhat more difficult to interpret given PRE-POST variations in whole-body movement patterns observed (possibly due to fatigue- or learning-related changes in coordination and control) and the potential effects of muscle fatigue or skin temperature on EMG signal amplitudes, muscle activation levels were significantly different in 4 muscle groups (RRA, LRA, LLD, LLES) between the PRE-POST trials ($p \leq 0.0487$) despite producing equivalent L4/L5 moments over time ($p \geq 0.6472$). Lumbar motion was consistent between the PRE and POST trial conditions ($p \geq 0.1212$).

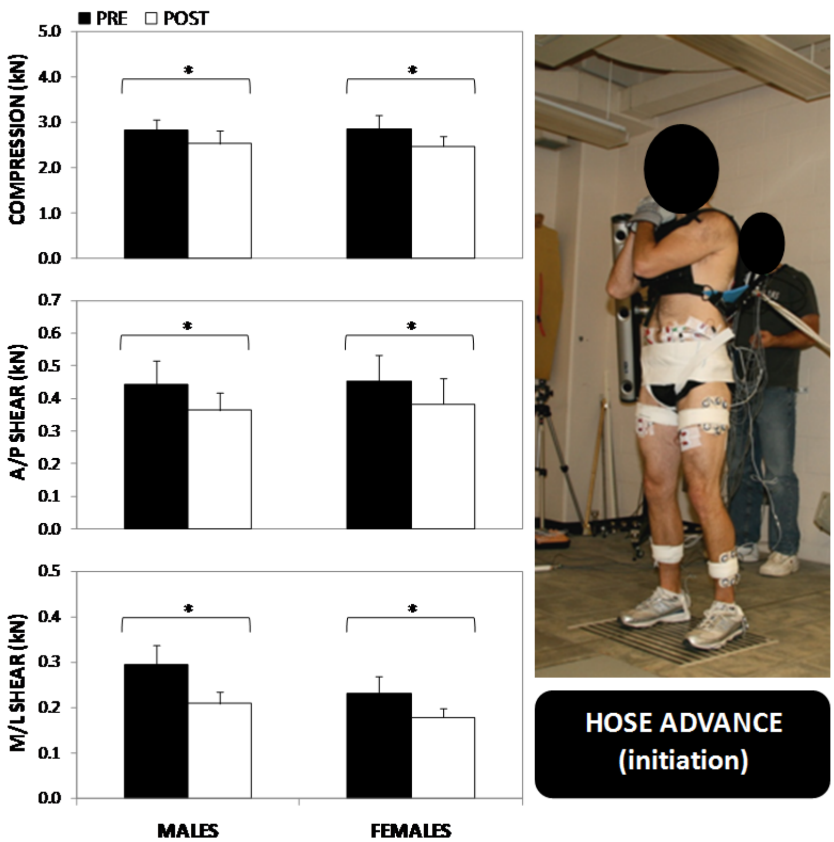

Figure 3.12. Peak low-back loading demands during the initiation phase of the Hose Advance task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Hose Advance (stride)

Different from what was observed in the initiation of the Hose Advance task, peak L4/L5 joint forces in males were 34.9% (1046 N), 48.8% (230 N), and 53.2% (105 N) greater in the compression (p = 0.0017), A/P shear (p = 0.0366), and M/L shear (p = 0.0027) directions, respectively (Figure 3.13). This was again related to gender-based differences in gain factors (Table 3.7). However, in this case, male-female differences were detected in net L4/L5 moments ($p \leq 0.0279$) without corresponding gender-based differences in trunk muscle activation levels ($p \geq 0.4384$); females produced lower magnitude L4/L5 moments than males despite using the same levels of trunk muscle activation and 8° less lumbar twist motion than men (p = 0.0003).

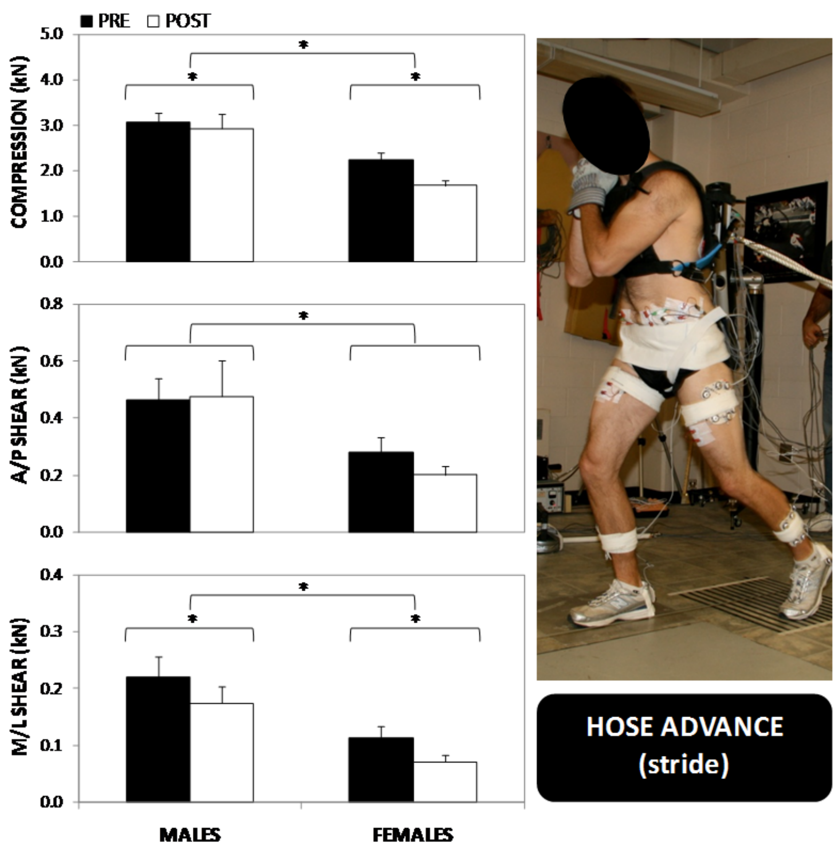

Figure 3.13. Peak low-back loading demands during the stride phase of the Hose Advance task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

In comparison to those measured during the stride component of PRE trials, peak L4/L5 forces were 13.3% (352 N) lower in compression (p = 0.0067) and 26.8% (45 N) lower in M/L shear (p = 0.0470) during POST trials (Figure 3.13). This was again associated with differences in PRE-POST gain factors (Table 3.7), driven this time by POST trial reductions in peak L4/L5 flexion moments (p = 0.0021) with only a small (≈ 2°), though statistically significant PRE-to-POST decrease in lumbar twist motion (p = 0.0326). Likely related to variations in inter-individual responses, group peak A/P shear forces were not different between the PRE and POST conditions (p = 0.4570).

Kneeling Hose Pull

When pulling a rope (hand-over-hand) from a half-kneeling position, females produced significantly lower magnitude L4/L5 extension (p = 0.0196) and lateral bend (p = 0.0102) moments than males despite exhibiting significantly greater ($p \leq 0.0419$) activation levels in 4 muscles (RLD, LRA, LUES, LLES). No gender-based differences were observed in lumbar motion ($p \geq 0.2741$). Again attributed to these findings was that gain factor magnitudes were significantly lower in females (Table 3.7) and peak L4/L5 compression (p = 0.0196) and A/P shear (p = 0.0458) forces were 20% (733 N) and 29.5% (275 N) lower in females as a result (Figure 3.14).

In comparison to the PRE trials, subjects produced relatively small (≈ 10 N•m) increases in L4/L5 twist moment magnitudes (p = 0.0147) in POST trials while tending to exhibit greater activation levels ($p \leq 0.0669$) in 9 muscles (RIO, RLD, RUES, RLES, LRA LEO, LLD, LUES, LLES). Given the nearly uniform increase in trunk muscle activity with only minor changes in L4/L5 moment magnitudes, it was not surprising that gain factors were significantly smaller in POST trials (Table 3.7). Somewhat more difficult to explain was that while PRE-POST differences were documented in gain factor magnitudes, there were no corresponding differences in peak L4/L5 joint forces ($p \geq 0.3558$) (Figure 3.14). Though only speculative, it is possible that this discrepancy was related to the fact that subjects used 12° more lumbar twist motion (p = 0.0020) and 5° more lateral bend motion (p = 0.0310) in POST trials. Specifically, the force- and moment-producing capabilities of trunk muscles could have been compromised in POST trials due to muscle elongation, fatigue, or changes in the muscle lines-of-action; if so, greater levels of muscle activation would have been required to produce L4/L5 moments that were comparable to those in PRE trials.

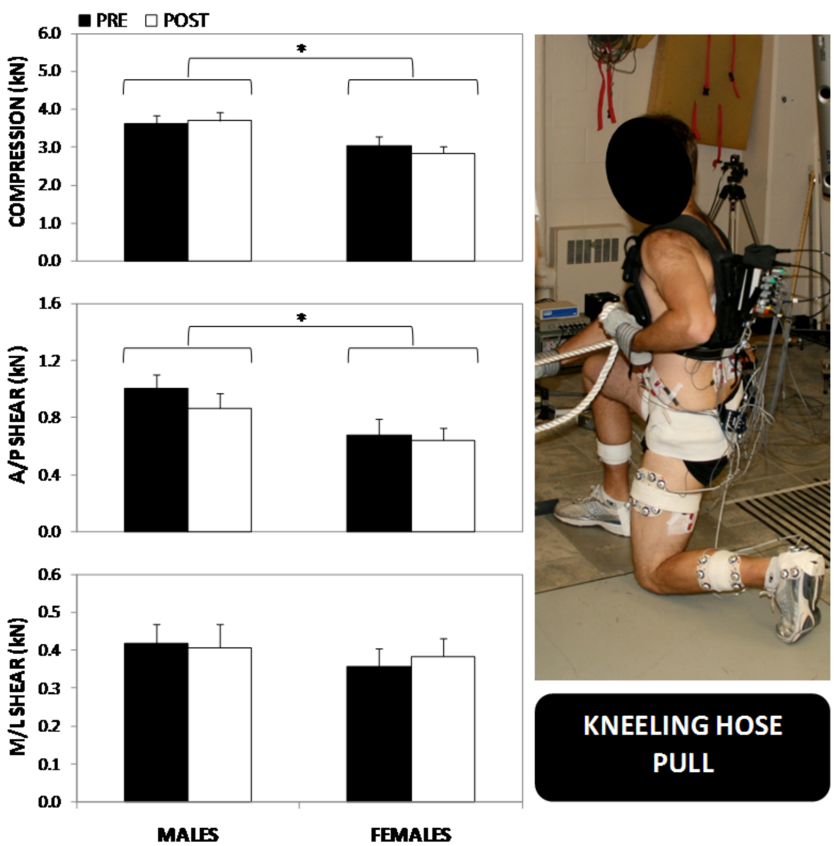

Figure 3.14. Peak low-back loading demands during the Kneeling Hose Pull task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant (p < 0.05).

Equipment Lift

In comparison to males, females again tended to use greater trunk muscle activation to produce smaller L4/L5 extension ($p < 0.0001$) and lateral bend ($p < 0.0001$) moments when performing the Equipment Lift task. Gender-based differences in activation levels were statistically significant ($p \leq 0.0318$) in only 2 muscles (LRA, LUES), but for all but 3 muscle groups (RRA, REO, LLES), peak activation levels were over 18% higher in women than in men. No gender-based differences in lumbar spine motion were observed ($p \geq 0.1245$). These findings were again reflected in gain factor magnitudes (Table 3.7) and translated into females experiencing peak L4/L5 joint compression ($p = 0.0014$), A/P shear ($p = 0.0006$), and M/L shear ($p = 0.0494$) forces that were, respectively, 24.3% (1390 N), 45.3% (766 N), and 23.6% (63 N) smaller than those in the males (Figure 3.15).

When performing this task after stair-climbing, activation levels of 6 muscles (REO, RLD, RUES, RLES, LUES, LLES) were significantly greater ($p \leq 0.0145$) than those measured in the PRE trials. When combined with the finding that subjects produced larger magnitude L4/L5 extension moments ($p = 0.0286$) and smaller magnitude lateral bend moments ($p = 0.0013$) during POST trials, a slightly more complicated spine loading response emerged. Specifically, peak L4/L5 joint A/P shear forces were 13.4% (188 N) lower ($p = 0.0105$) in POST trials without corresponding PRE-POST differences in compression ($p = 0.2409$) or M/L shear ($p = 0.3874$) forces (Figure 3.15). Although gain factor values were of lower magnitudes in POST trials (Table 3.7) and lumbar motion was not different between PRE and POST conditions ($p = 0.1941$), PRE-POST differences in gain factors could not be directly linked to PRE-POST differences in trunk muscle activity given that L4/L5 moment magnitudes differed in an inconsistent way. Accordingly, a less predictable link between gain factor values and peak L4/L5 joint forces would be expected.

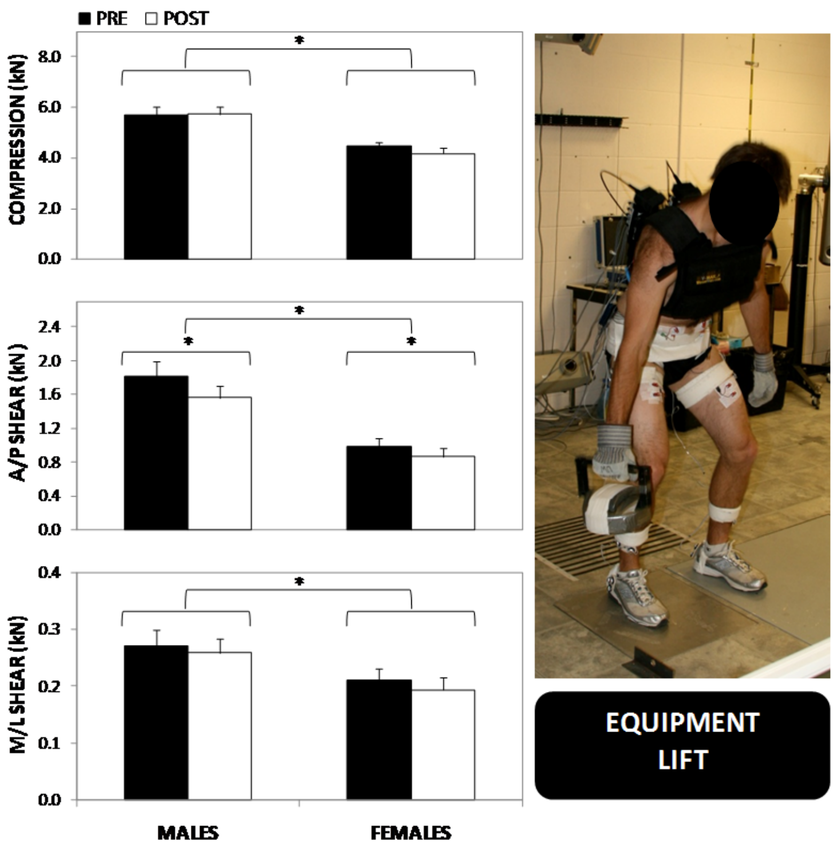

Figure 3.15. Peak low-back loading demands during the Equipment Lift task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Equipment Carry

There were no gender-based differences in trunk muscle activation levels ($p \geq 0.1258$), but L4/L5 joint moments were of significantly greater magnitudes in males ($p \leq 0.0326$). Additionally, females exhibited 1° and 6° less lumbar motion than did males about the lateral bend ($p = 0.0106$) and twist ($p < 0.0001$) axes, respectively . The net effect of these findings were that peak L4/L5 joint compression forces were 26.8% (682 N) lower in females ($p = 0.0047$), while peak A/P shear ($p = 0.1441$) and M/L shear ($p = 0.7681$) forces were not different between males and females (Figure 3.16). The lack of significant differences in peak L4/L5 joint forces, despite the existence of gender-based differences in gain factors (Table 3.7), was likely because muscular contributions to total joint shear forces were relatively small given the body posture and the task demand. (i.e., external load was axially applied to a vertically oriented trunk with a near-neutral lumbar spine posture [peak deviations about any lumbar axis did not exceed 8°]).

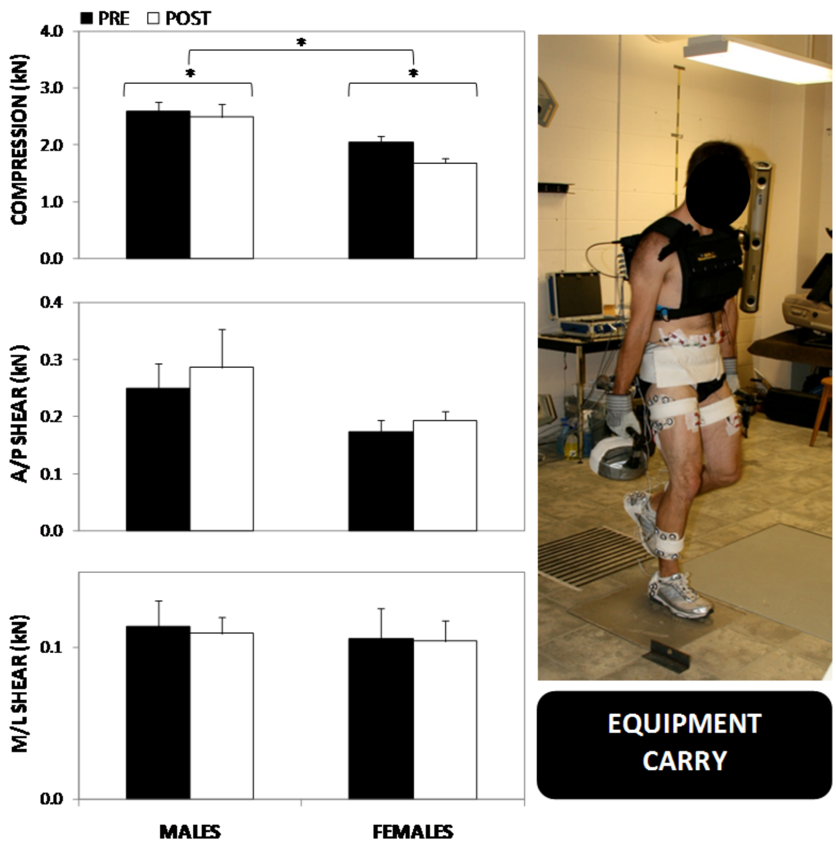

Figure 3.16. Peak low-back loading demands during the Equipment Carry task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Given that body positioning was relatively constrained given the objectives of this task, no PRE-POST L4/L5 joint moments ($p \geq 0.4228$) were observed and only a small (≈ 2°), though statistically significant decrease in lumbar twist motion was detected. However, significant increases ($p \leq 0.0428$) in the POST trial activation levels of 4 muscles (RLES, LLD, LUES, LLES) resulted in the calculation of lower magnitude POST trial gain factors (Table 3.7). Decreases of 9.9% (229 N) in peak L4/L5 joint compression forces ($p = 0.0050$) were likely influenced by the change in gain factor values, but PRE-POST peak L4/L5 shear force magnitudes were unaffected ($p \geq 0.1441$) given the relatively minor muscular contributions to total joint shear forces in this task (Figure 3.16).

Forcible Entry

Peak L4/L5 joint compression ($p = 0.0395$), A/P shear ($p = 0.0157$), and M/L shear ($p = 0.0051$) were, respectively, 29.5% (2103 N), 35.1% (434 N), and 38.5% (273 N) lower in females than in males (Figure 3.17). No gender-based differences were detected in lumbar motion ($p \geq 0.2606$), but L4/L5 twist moment magnitudes were significantly greater in male subjects ($p < 0.0001$). Because trunk muscle activation levels were not different between males and females ($p \geq 0.1182$), the gender differences in spine loading were again consistent with what would be expected based on gain factor values (Table 3.7).

During POST trials, peak L4/L5 joint compression forces were 11.1% (713 N) lower that those calculated during PRE trials ($p = 0.0161$), but there were no statistically significant PRE-POST differences in peak L4/L5 joint shear forces ($p \geq 0.1037$) (Figure 3.17). Though not statistically significant ($p = 0.0578$), POST trial gain factor values tended to be lower than those calculated in the PRE trials (Table 3.7) and again likely influenced the L4/L5 joint compression force magnitudes. In the POST trials, subjects exhibited a small increase (≈ 3°) in lumbar twist motion ($p = 0.0464$), but no differences were discovered in PRE-POST L4/L5 moment magnitudes ($p \geq 0.2024$). Since similar PRE-POST L4/L5 moments were produced with relatively minor PRE-POST differences in lumbar postures, it was somewhat unexpected to find that POST trial muscle activation levels tended to be lower ($0.0506 \leq p \leq 0.0917$) in 5 muscles (RRA, LRA, LEO, LIO, LLD) given that lower magnitude POST trial gain factor values were calculated (Table 3.7). However, POST trial activation levels were significantly higher ($p \leq 0.0065$) in 2 muscles (RLES, LLES) and tended to be higher ($p = 0.0679$) in one other muscle (RUES). This POST trial muscle synergy resulted in larger magnitude internal L4/L5 moments than did the PRE trial muscle synergy, and the POST trial gain factor values tended to be lower as a consequence.

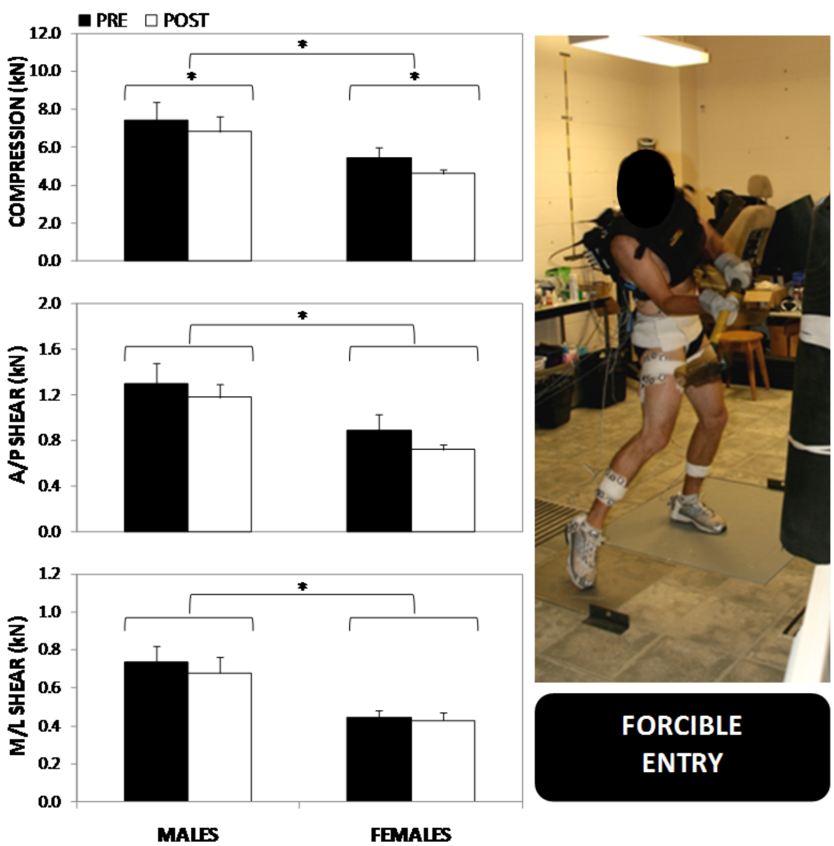

Figure 3.17. Peak low-back loading demands during the Forcible Entry task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Victim Rescue (drag)

Despite using similar lumbar motion ($p \geq 0.1872$) and producing lower magnitude L4/L5 extension moments ($p = 0.0190$), females exhibited significantly greater activation levels ($p \leq 0.0341$) in 4 muscles (REO, RLES, LRA, LEO) and activation levels that tended to be higher ($0.0565 \leq p \leq 0.0714$) in 3 more muscles (LIO, LUES, LLES). As would be expected, female gain factor values were thus of lower magnitudes than those for the males (Table 3.7) and peak L4/L5 joint compression ($p = 0.0086$) and A/P shear ($p = 0.0177$) forces were, respectively, 22.7% (929 N) and 44.4% (370 N) lower in females as a result (Figure 3.18). Peak L4/L5 joint M/L shear forces were unexpectedly greater in females ($p = 0.0398$); however, the magnitude of the male-female difference (51 N) was likely biomechanically insignificant.

In POST trials, subjects produced significantly smaller magnitude L4/L5 extension moments ($p = 0.0002$) using trunk muscle activation levels that were not different ($p \geq 0.1247$) from those measured in PRE trials. Since no PRE-POST differences in lumbar motion were observed ($p \geq 0.0909$), it was again explicable that POST trial gain factor magnitudes were lower than the PRE trial values (Table 3.7). As a result, peak L4/L5 joint compression ($p = 0.0077$) and A/P shear ($p = 0.0178$) forces, were respectively, 9.6% (365 N) and 21.4% (156 N) lower in POST trials (Figure 3.18). No PRE-POST difference was found in peak L4/L5 joint M/L shear forces were found ($p = 0.5008$), likely because low-back task demands were confined mainly to the sagittal plane.

Victim Rescue (initiation)

Females produced lower magnitude L4/L5 extension ($p < 0.0001$) and lateral bend ($p < 0.0001$) moments than males despite exhibiting no differences in lumbar flexion ($p = 0.6697$) or lateral bend ($p =$

0.3352) motion and again using significantly greater levels of activation ($p \leq 0.0413$) in 6 muscles (REO, RLD, LRA, LIO, LLD, LUES). As before, this caused gain factors values to be lower in females (Table 3.7) and peak L4/L5 joint compression ($p = 0.0013$), A/P shear ($p = 0.0138$), and M/L shear ($p = 0.0411$) forces were, respectively, 26.5% (1603 N), 42.4% (542 N), and 15.4% (24 N) lower in females as a result (Figure 3.19). Though lumbar twist motion tended to be greater in males, the magnitude of the difference (< 1°) was not likely biomechanically relevant nor was it statistically significant ($p = 0.0612$).

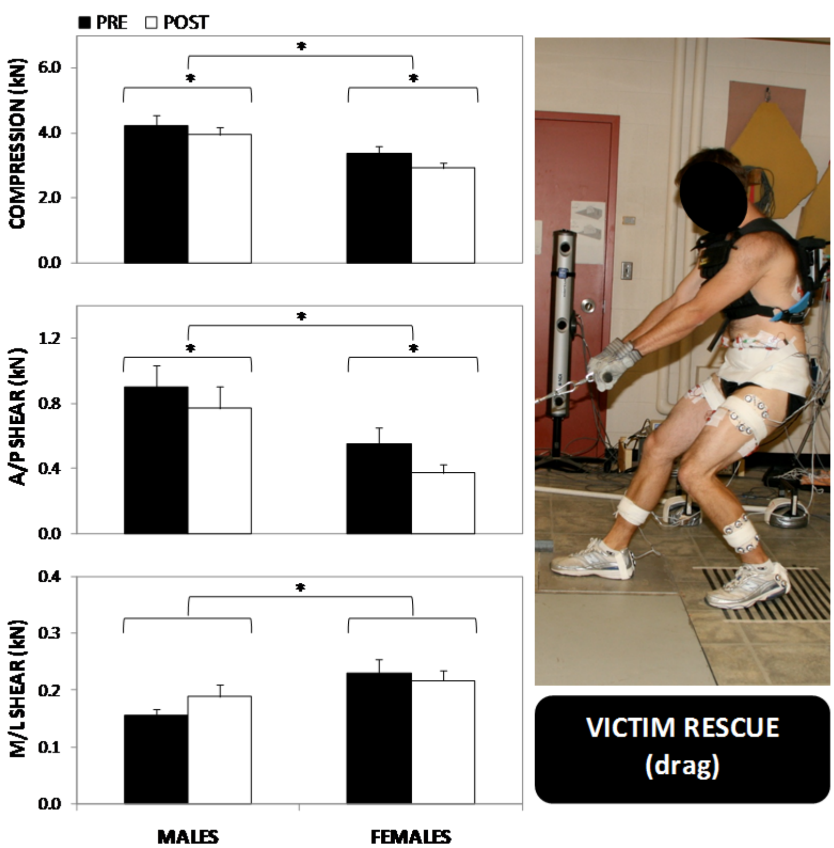

Figure 3.18. Peak low-back loading demands during the drag phase of the Victim Rescue task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Different from what was found in the above tasks, there were no PRE-POST differences in gain factor values (Table 3.7) nor were there statistically significant PRE-POST differences in peak L4/L5 joint force magnitudes ($p \geq 0.0770$) (Figure 3.19). This was not surprising given that PRE-POST trunk muscle activation levels were not significantly different ($p \geq 0.0760$) and the only difference in L4/L5 moment magnitude was a small (≈ 3 N•m), though statistically significant increase about the twist axis (p = 0.0045). Lumbar flexion (p = 0.0174), lateral bend (p = 0.0131), and twist (p = 0.0341) motions were, respectively, 6°, 1°, and 1° greater in the POST trials. Although PRE-POST differences in lumbar motions were of small magnitudes, it is possible that the differences impacted gain factor values for reasons discussed above.

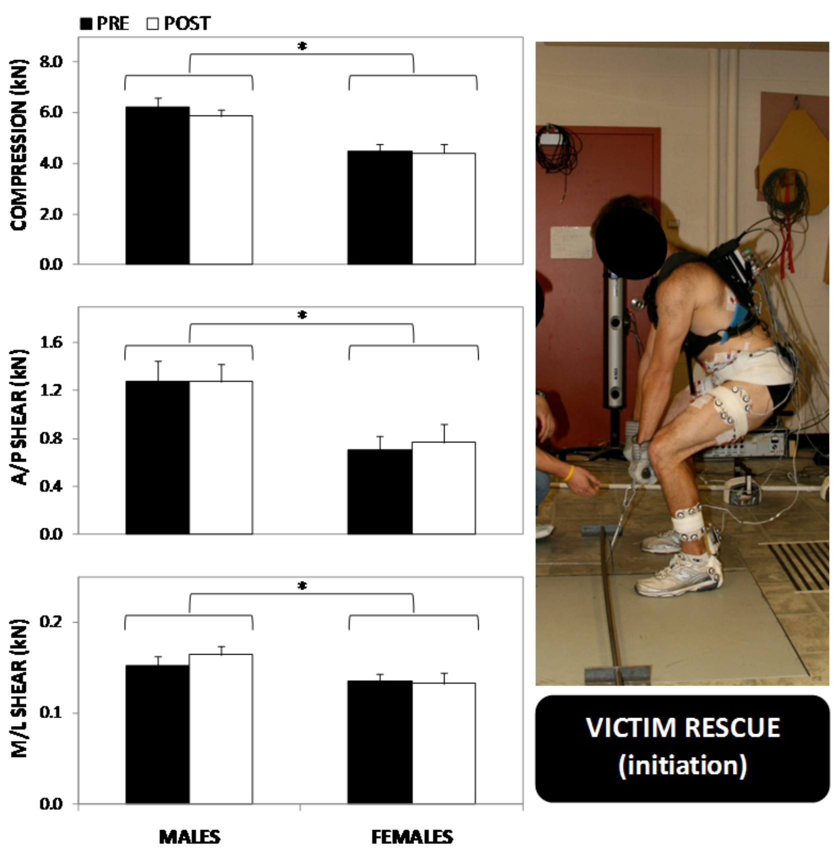

Figure 3.19. Peak low-back loading demands during the initiation phase of the Victim Rescue task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Ceiling Breach

In comparison to those recorded in male subjects, activation levels of 3 trunk muscles (RUES, RLES, LRA) were significantly greater in females ($p \leq 0.0393$); a similar trend ($0.0593 \leq p \leq 0.0758$) was observed in 2 other muscles (LLES, LEO). In addition to these disparities, females exhibited 9° more lumbar flexion motion than did males ($p = 0.0153$). Despite these gender-based differences, L4/L5 moment magnitudes were not different between male and female subjects ($p \geq 0.1699$). Again related to the fact the female subjects exhibited greater trunk muscle activation levels to generate L4/L5 moments of magnitudes to those of males, gain factor values were significantly lower for females (Table 3.7). The net effect of these findings was that peak L4/L5 joint compression forces were 24.4% (1238 N) lower in females ($p = 0.0254$); no gender-based differences were observed in peak L4/L5 joint A/P shear ($p = 0.3735$) or M/L shear ($p = 0.1856$) forces (Figure 3.20).

There were no PRE-POST differences in lumbar spine motion ($p \geq 0.1122$) or L4/L5 moment magnitudes ($p \geq 0.1010$), and in all but one muscle group (LLES, $p = 0.0343$), activation levels were not significantly different between PRE and POST trials ($p \geq 0.1212$). As a result, there were no PRE-POST differences in gain factor values (Table 3.7) or peak L4/L5 joint forces ($p \geq 0.2505$) (Figure 3.20).

Ceiling Pull

Peak L4/L5 joint compression ($p = 0.0003$), A/P shear ($p = 0.0193$), and M/L shear ($p = 0.0155$) forces were, respectively, 32.8% (1289 N), 42% (269 N), and 33.8% (92 N) lower in females than in males (Figure 3.21). This could be largely attributed to gender-based differences in gain factor values (Table 3.7), which resulted because males generated higher magnitude L4/L5 flexion moments ($p = 0.0046$) despite exhibiting trunk muscle activation levels that were, with only one exception (RLES, $p = 0.0428$), not different from those of females ($p \geq 0.1300$). However, the gain factor could have also been influenced by the finding that females exhibited 12° and 6° more lumbar twist ($p = 0.0034$) and lateral bend ($p = 0.0010$) motion than males, respectively.

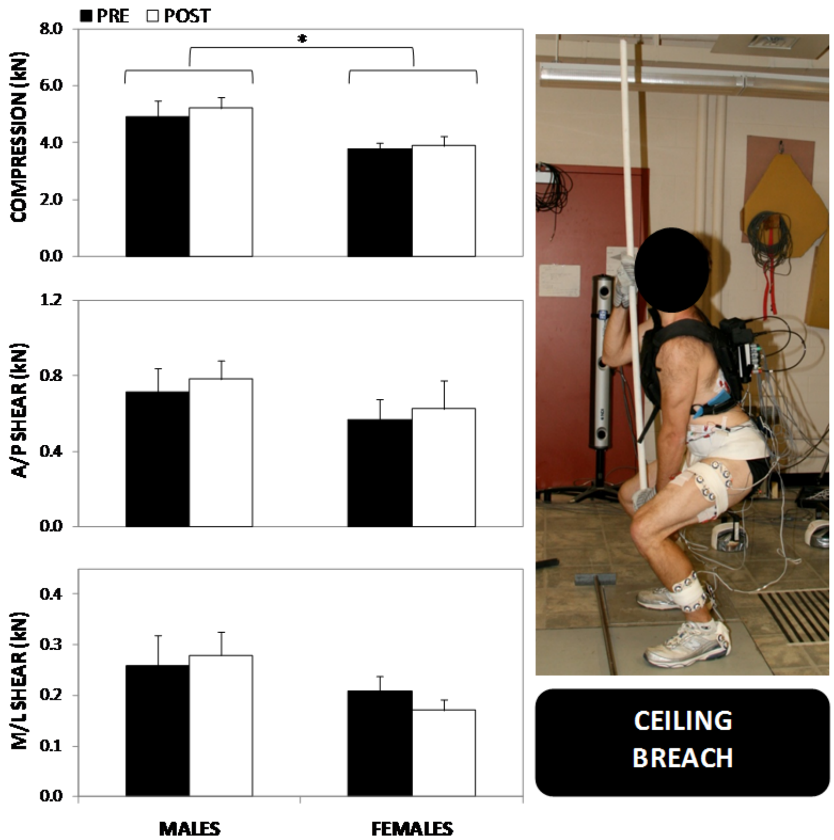

Figure 3.20. Peak low-back loading demands during the Ceiling Breach task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

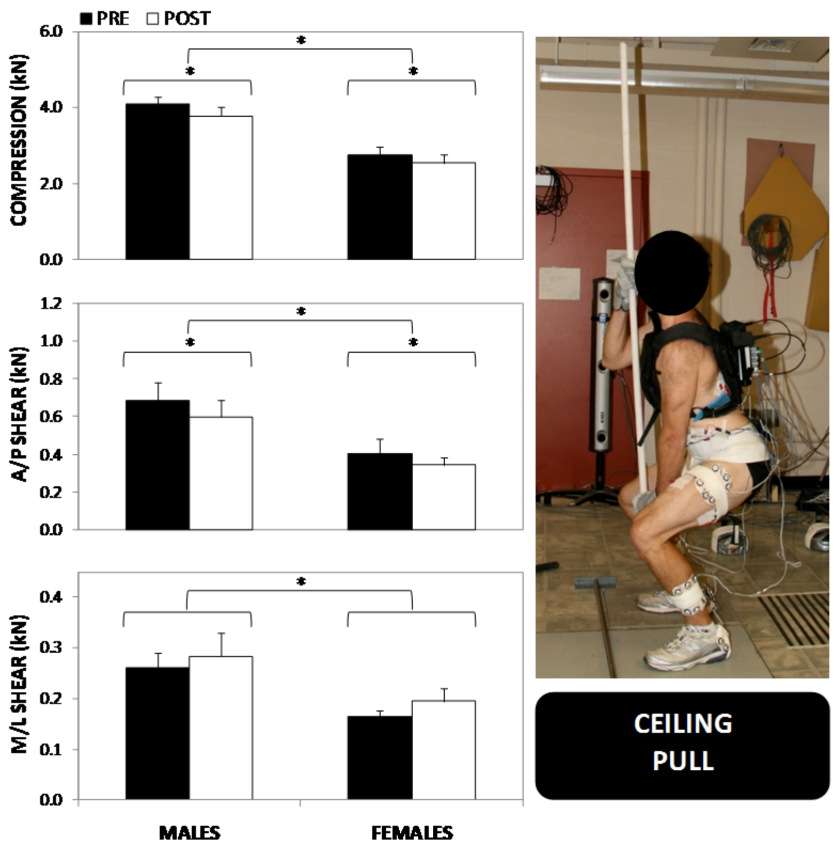

Figure 3.21. Peak low-back loading demands during the Ceiling Pull task. Mean values for all male (N = 10) and female (N = 10) subjects are reported; error bars represent the standard error of the mean. * indicates that differences between the means were statistically significant ($p < 0.05$).

Though no PRE-POST differences were observed in lumbar spine motion ($p \geq 0.1022$) or L4/L5 moment magnitudes ($p \geq 0.1393$), POST trial peak L4/L5 joint compression ($p = 0.0140$) and A/P shear ($p = 0.0657$) shear forces were, respectively, 7.7% (262 N) and 13.5% (74 N) lower than those calculated in PRE trials (Figure 3.21). Supported by the fact that PRE-POST gain factor values were not different (Table 3.7), it could be argued that the magnitudes of PRE-POST differences in peak L4/L5 joint forces were biomechanically minor. However, it is important to report that the minor PRE-POST differences in peak L4/L5 joint forces resulted notwithstanding decreases ($p \leq 0.0056$) in the POST trial activation levels of 2 muscles (RRA, RLD) and increases ($p \leq 0.0316$) in 2 others (RUES, LLES).

3.4. Discussion

Firefighting tasks are typically regarded as “heavy”, “forceful”, “strenuous”, or “arduous”, but such descriptors may be ambiguous from a biomechanical standpoint given that personal movement strategies can alter the internal joint loading response to fixed external task and environmental constraints (McGill 2004). In this study, an EMG-driven musculoskeletal modeling approach that was sensitive to intra- and inter-individual movement and muscle activation patterns was used to quantify peak low-back compression and shear loading demands during laboratory-simulated firefighting tasks and to examine the impact of fatigue on the peak low-back loading response. Attributable to inter-individual differences in body sizes, trunk muscle activation levels, and movement patterns was the finding that peak L4/L5 joint forces varied considerably between subjects. Not surprisingly, between-task differences in low-back loading demands were also considerable. Fatigue-related changes in trunk muscle activation levels and movement patterns resulted in intra-individual variations in low-back loading demands. Overall, results suggest that physical characteristics of individuals, tasks performed, and fatigue can influence low-back loading demands and injury potential in firefighters.

Interpretation and implications of the study findings are dependent on the outputs of a musculoskeletal model, and it is therefore helpful when interpreting the results to consider the impact of the general modeling approach employed and the inherent assumptions therein. In anticipation of this and given that peak low-back load magnitudes are directly related to the amount of “adjustment” made to EMG-based trunk muscle force estimates, the magnitude of the common gain factor (*G* in the Methods section) was used to guide in the Results section to interpret the low-back loading outcomes. Since whole-body movement and trunk muscle activations varied in an inconsistent and complex manner between subjects, tasks, and over time, it was not possible to identify a single or consistent physiological reason why gain factor magnitudes varied as they did, but general trends did emerge and are expanded on below as a way to interpret the results and consider their implications.

No attempt was made in this study to scale the anatomical description in the musculoskeletal model to account for inter-individual differences in the physiological cross-sectional areas, moment arms, or lines-of-action of ligaments or muscles. Model outputs are highly-sensitive to such parameters (Dieën and Looze 1999), as they directly impact estimates of the force- and moment-producing capacity of modeled tissues. Although subject-specific common gain factors were computed to individualize (calibrate) the model, this procedure is admittedly limited in its ability to accommodate inter-individual differences that might exist in the morphology and mechanical function of modeled tissues. This could partially explain why some gender-based differences in low-back loading remained even after peak magnitudes were normalized with respect to subject bodyweight. If the lumbar torsos of females were simply scaled-down versions of their male counterparts, then normalizing spinal load magnitudes by bodyweight would be expected to mitigate most gender differences (Marras et al. 2002; Marras et al. 2003). The fact that gain factor magnitudes were typically less than 1.0 in females and greater than 1.0 in males suggest that some of the between-gender variation was likely attributed to male-female differences in body size, especially given that few gender differences were noted in lumbar postures. Specifically, in cases where males and females were required to generate equivalent low-back moments (i.e., when task demands predominantly dictated external moment characteristics) while also maintaining sufficient spinal stiffness, greater muscle co-activity would have been needed by females to compensate for their smaller muscles and moment-arms (Marras et al. 2001b). However, perhaps due to sexual dimorphism in the pelvis and thorax (Patriquin et al. 2003; Tague 2005), Jorgensen, Marras, and colleagues (Jorgensen et al. 2001; Marras et al. 2001b) have shown that trunk muscle lines-of-action also vary considerably between men and women and demonstrated that gender differences in the low-back loading demands are not solely due to the fact that men are larger than women (Marras et al.

2002; Marras et al. 2003). These gender differences in trunk muscle lines-of-action could also partially explain why males and females exhibit different spine stabilization strategies (Granata and Orishimo 2001) and why male-female muscle activation patterns do not similarly correspond to those predicted using optimization-assisted musculoskeletal models that do not explicitly account for gender differences in muscle lines-of-action (McMulkin et al. 2003). In light of these findings, it is not surprising that gender differences in peak low-back loading demands persisted in this study even after bodyweight normalization, as muscle activation patterns would be expected to differ given inherent gender differences in musculoskeletal geometry. But what needs to be emphasized is that because the approach taken to individualize the musculoskeletal model was limited to the computation of subject-specific gain factors, female low-back loading demands and injury potential could have been underestimated in this study.

Low-back loading demands varied considerably between subjects, and it is possible that some of this variability was due to modeling limitations and assumptions described above. However, another way to interpret the between-subject variability is that some individuals employed movement strategies that resulted in superfluous peak spinal load magnitudes. The musculoskeletal linkage is endowed with numerous biomechanical degrees-of-freedom and thus individuals could conceivably draw from a pool of many possible movement strategies to successfully meet task objectives. At any given point in time, a number of interacting endogenous and exogenous factors can influence the movement strategy employed (Davids et al. 2003; Glazier and Davids 2009). Though it was not deemed practical to delineate all potential causes of inter-individual variation due to the complex nature and numerous sources of data used to estimate these quantities (the study was not designed to specifically address this question), results of crude computations made (i.e., bodyweight normalization of peak spinal loads) suggested that between-subject differences in peak low-back loading demands were not solely due to differences in body sizes. This interpretation was further supported by the observation that whole-body movement and trunk muscle activation patterns were variable between subjects, despite performing relatively constrained tasks within a controlled laboratory environment. Similar observations have been reported previously (Granata et al. 1999; Dieën et al. 2001). Accordingly, since it is often impossible or impractical to alter the physical demands fireground tasks through task modification, it might be possible to devise worker-focused strategies that incorporate movement-focused education and exercise programs which aim to attenuate peak spinal loads through the modification of habitual movement behaviours.

If it is accepted that the simulated tasks sufficiently represented the external physical demands associated with basic firefighting duties, results of this study indicate that performing such tasks could be hazardous for some people even if personal movement strategies can modulate low-back loading demands. In all tasks examined, recommended action limits for peak lumbar compression and A/P shear forces were exceeded in at least one subject. In all but one task (Equipment Carry), maximum permissible limits for peak compression or A/P shear forces were also exceeded at least once. Though it could be argued that such population-based low-back injury prevention guidelines may be overly conservative if used to protect what many assume to represent a "typical" firefighter or candidate (i.e., young, large, physically fit, male), the lumbar spines of incumbents who are older, smaller (e.g., females), physically unfit, or who have a low-back injury history could be particularly vulnerable to damage if exposed to forces in excess of recommended limits. Clearly, subjects in this study comprised a relatively homogenous group of young, pain-free individuals with no experience in firefighting, and caution should be exercised if attempting to extrapolate study findings for applications involving incumbent firefighters. However, because all subjects were able to pass the CPAT, results are at the very least relevant to firefighter candidates or new hires who may have minimal or no previous training and who are without a low-back injury history. Moreover, results of this study suggest that performing the CPAT itself may expose some individuals to potentially hazardous low-back loading levels. Given the frequently held opinion that firefighting tasks are inherently hazardous, these findings were not unexpected. Nevertheless, data from this study can be used as a guide in future intervention efforts aimed at attenuating peak low-back loading demands using either engineering or administrative ergonomic controls.

Somewhat more difficult to interpret was the finding that low-back loading demands tended to decrease following the stair-climbing protocol, a response again largely driven (computationally) by the direction of change in gain factor magnitude. However, there are a number of complicating factors and assumptions that must be acknowledged before concluding that fatigued firefighters might be less likely to sustain low-back injuries. First, an objective measure of fatigue was not made, and it was thus possible that subjects chose to terminate data collection sessions for reasons unrelated to fatigue (e.g., lack of motivation). Consequently, PRE-POST trial differences in peak spine loading responses of some individuals may not have been fatigue-related. For example, although subjects were provided with an opportunity to practice the tasks before data were collected, it is possible that their movement strategies changed as they acclimated and became more comfortable in the laboratory.

A second and arguably more important factor to consider when attempting to interpret PRE-POST trial differences detected in low-back loading demands is that clear and consistent explanations were not found to account for why gain factor magnitudes changed as reported and how these changes impacted loading estimates. Due to the unique morphological, physiological, and psychological make-up of study subjects, it is likely that clear and consistent explanations for PRE-POST trial differences were indefinable because movement strategy adaptations varied between subjects. It was observed, for example, that some subjects used uniformly greater muscle activity in the POST trials to generate equivalent low-back moments to generate equivalent moments; this increase in trunk muscle co-activity would have caused gain factor magnitudes to decrease and its impact on spinal loading may been underestimated based on the method of gain computation employed. Increased co-activity might have been utilized to compensate for fatigued trunk muscles in the control of spinal posture, motion, and moment generation (Granata et al. 2004; Grondin and Potvin 2009; O'Brien and Potvin 1997; Potvin and O'Brien 1998; Sparto and Parnianpour 1999). Alternatively, POST trial decreases in gain factor magnitudes were attributed in some cases when muscle co-activity decreased (i.e., decreased antagonistic muscle contraction), perhaps to conserve metabolic energy and physiological work capacity. When this occurred, as it did in a study by Gregory et al. (2008), low-back loading demands decreased following stair-climbing, but the stability of the spinal system may have been compromised (Granata et al. 2004). In either case (i.e., increased or decreased trunk muscle co-activation), apparent fatigue-related reductions in low-back loading demands may not necessarily indicate that the low-back injury potential of firefighters would be reduced when fatigued. It is especially important to appreciate that the low-back loading capacity could decrease during fatiguing fireground operations, and thus lower-magnitude loading demands could still lead to injury. Of course, it is also important to acknowledge that changes in EMG signal amplitudes could have also have been an artifact due to changes in skin temperature and/or conductive properties at the skin-electrode contact surface.

A third factor to consider when comparing the PRE and POST trial low-back loading demands is that there were no differences in the PRE-POST gain factor magnitudes computed for the last three tasks in the circuit (Victim Rescue, Ceiling Breach, Ceiling Pull), suggesting that subjects may have experienced some amount of recovery. If not restricted by laboratory space and instrumentation requirements, it would have been possible to control for the potential influence of subject recovery by randomizing task presentation, but it was not practical to alter the order in which tasks were performed given the aforementioned experimental constraints.

Limitations

The general modeling approach employed has been well-documented in publications spanning nearly three decades (Davis and Jorgensen 2005; Reeves and Cholewicki 2003) and several modeling-related limitations were highlighted above. However, issues related to the validity of the approach remain unresolved. Without being able to measure tissue forces directly, it still not possible to determine if the mathematically-justified optimization-assisted multi-tissue musculoskeletal models yield tissue force estimates that more closely correspond to reality than do those generated by the more physiologically-justified EMG-assisted models. Recent evidence suggests that if the goal is to use such models for relative comparisons (e.g., comparing net low-back loading demands between experimental conditions), the choice of approach may be made based on preference or practical considerations (Dieën and Kingma 2005). Nevertheless, given the objectives of this study, the modeling approach employed was justified based on the fact that estimates of low-back loading demands were sensitive to inter- and intra-individual variations in movement strategies, a general limitation of the optimization-assisted approach (Reeves and Cholewicki 2003).

3.5. Conclusions

Results of this study demonstrate that physical characteristics of individuals, duties performed, and fatigue can influence low-back loading demands and injury potential in firefighters. However, it is important to emphasize that despite variations in the peak spinal load response between subjects, tasks, and over time, recommended acute exposure limits were routinely exceeded under the conditions examined. Therefore, this study confirms that firefighter tasks are inherently hazardous for low-back health and that administrative controls (e.g., pre-placement screening and training) remain critical elements in low-back injury prevention and rehabilitation strategies for firefighters, especially given that many job duties are non-modifiable and unpredictable.

CHAPTER 4

Ankle Immobilization alters Lifting Kinematics and Kinetics – Occupational Low-Back Loading Demands and Potential for Injury

CHAPTER 4

Ankle Immobilization alters Lifting Kinematics and Kinetics – Occupational Low-Back Loading Demands and Potential for Injury

Summary

Background: Theoretical and empirical data support the notion that distal lower extremity joint dysfunction could influence the low-back injury potential of workers. The objective of this experiment was to examine the influence of unilateral ankle immobilization on the kinematics and kinetics of lifting.

Methods: With and without their right ankle immobilized, 10 male volunteers performed laboratory-simulated occupational lifting tasks. Together with force platform data, three-dimensional kinematics of the lumbar spine, pelvis, and lower extremities were collected, and a dynamic biomechanical model was used to calculate peak compressive and shear loads imposed on the L4/L5 intervertebral joint.

Results: In comparison to the unaffected conditions, ankle immobilization resulted in less knee ($0.0004 \le p \le 0.0697$) and greater lumbar spine ($0.0006 \le p \le 0.3491$) sagittal motion when lifting. Associated with this compensatory movement strategy were greater L4/L5 anterior/posterior reaction shear forces ($0.0009 \le p \le 0.2450$). However, in a few cases where individual compensatory movement strategies differed from the "group" response (i.e., subjects increased their sagittal knee and hip motion on the affected side), peak L4/L5 joint compressive loads increased while the peak L4/L5 anterior-posterior did not change.

Conclusions: Distal lower extremity joint dysfunction can alter the way in which individuals move and load their low-backs when lifting. The specific ways in which individuals compensate for personal movement constraints could alter the potential site and mechanism of occupational low-back injury.

4.1. Introduction

For many scientific and practical applications, the human body is modeled as a series of interconnected segments. When the body is modeled this way (as a “kinematic chain”), fundamental principles of mechanics can be applied to demonstrate that motion and dynamics of the human movement system are inherently coupled (Zajac and Gordon 1989). It logically follows that injury- or treatment-induced anatomical movement constraints (e.g., joint immobilization) can alter the kinematics and kinetics of all body segments, especially when performing “closed” kinematic chain tasks such as lifting. Despite theoretical (Zajac 1993; Zajac et al. 2002; Nott et al. 2010) and empirical (DeLeo et al. 2004; Goodman et al. 2004; Radtka et al. 2006) support for the kinematic chain theory of human movement, occupational low-back injury prevention strategies often focus predominately on physical attributes of work tasks, systems, and environments with less emphasis placed on personal characteristics that could influence the way individuals move their bodies and load their low-back tissues when working. Given that low-back loading patterns and injury potential can be highly sensitive to personal movement strategies (McGill 2009; Reeves and Cholewicki 2003), efforts to identify personal characteristics that promote potentially injurious loading patterns are warranted. For physically demanding occupations such as firefighting and soldiering, where injuries are prevalent (Hauret et al. 2010; Karter and Molis 2010; Jennings et al. 2008; Reichard and Jackson 2010; Schneider 2001) and it is not typically feasible to modify work stations or tasks, knowledge of how personal characteristics influence movement strategies could be incorporated in the development of worker-focused interventions (e.g., exercise).

Injuries to the ankle and foot occur in various work environments (Grimm and Fallat 1999; Conti and Silverman 2002), but the number of distal extremity injuries amongst fire and military service personnel typically exceed what is reported in other jobs (Maguire et al. 2005). Injuries to the feet and ankles of these “occupational athletes” are often severe (e.g., fractures) (Knapik et al. 2003) and thus functional deficits (e.g., loss of ankle joint range-of-motion) can persist long after acute treatment (Faergemann et al. 1998; MacKenzie et al. 1993; Nightingale et al. 2007). Davis and Seol (2005) reported that individuals who had previous lower extremity injuries moved their lumbar spines differently than did their age-, gender-, and anthropometrically-matched uninjured counterparts when performing occupational lifting tasks. Interestingly, previously injured segments or joints most distal to the low-back were found to have a greater influence on lumbar spine kinematics than did injuries to more proximal segments or joints. Because low-back loading patterns and lumbar spine kinematics are intrinsically linked (Davis and Marras 2000a), it is possible that distal lower extremity joint dysfunction (e.g., injury- or treatment-induced deficits in ankle joint range-of-motion) could alter the potential for low-back injury when lifting at work. However, this notion has yet to be tested directly.

The purpose of this study was to determine what effect unilateral immobilization of the ankle joint would have on low-back loading while lifting. It was not the goal of this investigation to replicate a commonly encountered occupational scenario, but rather to gain insight into a specific situation that was observed when conducting a previous research project. Based on observations made during the parallel project, it was hypothesized that the peak low-back loading response to lifting would be influenced by unilateral ankle immobilization since individuals would be obliged to adapt their movement strategies to compensate for the loss of ankle joint range-of-motion.

4.2. Methods

Study Subjects

Ten male volunteers with no self-reported musculoskeletal pain or injuries were recruited from a university student population. Physical characteristics of subjects are summarized in Table 4.1. Subjects reported a mean score of 79.2 (SD = 1.8) out of 80 on the Lower Extremity Functional Scale (Binkley et al. 1999), indicating no significant functional limitations in the lower limbs. Upon arriving at the laboratory, subjects read and signed informed consent documentation that had been previously approved by the University of Waterloo's Office of Research Ethics.

Table 4.1. Height, mass, and age of 10 male study subjects.

Subject	Height (m)	Mass (kg)	Age (yrs)
S01	1.85	89.7	25
S02	1.93	81.5	29
S03	1.83	99.5	31
S04	1.83	80.1	28
S05	1.78	82.7	28
S06	1.96	79.0	20
S07	1.81	95.3	23
S08	1.93	99.8	22
S09	1.77	88.2	20
S10	1.79	74.0	22
Mean (SD)	1.85 (0.07)	87.0 (8.99)	24.8 (3.97)

Experimental Protocol

With and without their right ankle immobilized, subjects completed 18 permutations of a laboratory-simulated occupational lifting task. Specifically, from three different origins (Positions 1, 2, and 3 in Figure 4.1), a 3.7 kg and 12.7 kg mass was lifted to three different destinations (Positions 4, 5, and 6 in Figure 4.1). Lifting height was standardized (0.8 m) and three repetitions of each permutation were performed. In total, 108 lifts were performed by each subject (54 lifts in each ankle condition). Both the frequency and cycle time of each lift was self-selected, as subjects were instructed to take as much time as they deemed necessary between consecutive exertions to avoid perceiving fatigue. Order of task performance was randomized between subjects as was exposure to right ankle immobilization.

Immobilization of the right ankle was achieved through the use of a custom brace worn in each subject's own running shoes (Figure 4.2). The brace was constructed from lightweight materials (total mass of brace $\approx$ 0.2 kg) and was designed to restrict rotation of the right ankle in all three anatomical planes of motion during lifting. Before and after the brace was attached, subjects were asked to maximally rotate their ankle in the frontal, sagittal, and transverse planes with their foot planted firmly on the floor. This procedure was used to confirm that the bracing mechanism limited “closed kinematic chain” ankle motion without causing undue discomfort.

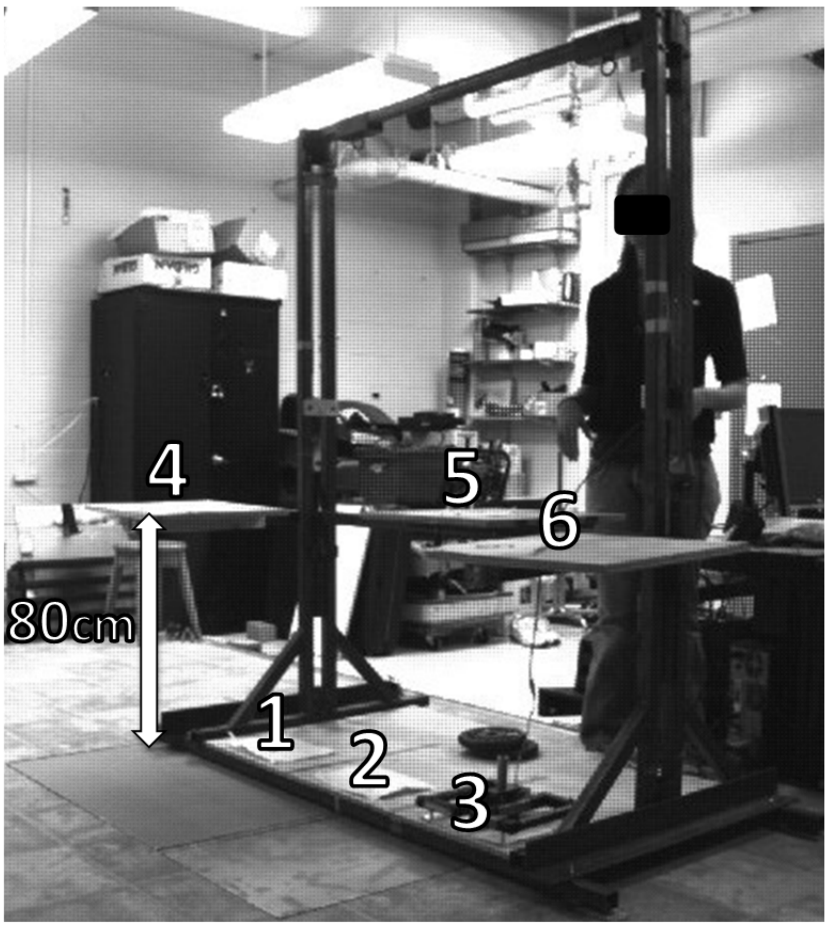

Figure 4.1. Physical configuration of the experimental set-up. Lift origins (labeled 1, 2, and 3), lift destinations (labeled 4, 5, and 6), and the lift height (80 cm) are defined. This configuration remained consistent across all subjects (N = 10).

Figure 4.2. A custom brace was applied as pictured to restrict ankle motion when lifting.

Data Acquisition

A custom lifting apparatus with handles was designed to allow weights to be added and removed between different lifting tasks. An analog switch was fastened to the bottom of the lifting apparatus to assist in the objective determination of lift initiation/termination events.

As described in Chapter 3, three-dimensional ground reaction forces were measured with two AMTI force platforms (Advanced Mechanical Technology, Inc., Watertown, MA, United States) and kinematics of the feet, shanks, thighs, pelvis and trunk were recorded with an Optotrak Certus® motion capture system (Northern Digital Inc., Waterloo, ON, Canada). Force platform and marker data were temporally synchronized with marker position data and sampled at rates of 1024 and 32 Hz, respectively, using an Optotrak® Data Acquisition Unit (ODAU II, Northern Digital Inc., Waterloo, ON, Canada).

Data Processing and Reduction

Exactly as described in Chapter 3, force platform data and kinematics of the lower body and trunk were input into Visual3D™ software (Version 4, C-Motion, Inc., Germantown, MD, United States) to construct a "bottom-up" three-dimensional inverse dynamical linked-segment model of the lower body and trunk. Orthogonal components of the three-dimensional L4/L5 net joint moment were then input into a third-order polynomial to calculate the L4/L5 joint compressive force while lifting. The polynomial was derived by McGill et al. (1996b) to estimate the L4/L5 joint ("bone-on-bone") compressive force given a three-dimensional net L4/L5 joint moment. No attempt was made in this investigation to quantify the L4/L5 bone-on-bone shear force, although anterior/posterior (A/P) and medial/lateral (M/L) shear components of L4/L5 reaction forces were calculated based on the inverse dynamics procedures described above. A bone-on-bone force refers to the joint contact force resulting from the net effects of all possible joint-loading sources (Winter 2005), whereas joint reaction forces resulting from inverse dynamics procedures represent only gravitational and inertial forces associated with moving body segments and externally applied forces or moments of forces (e.g., ground reaction force) (Zajac 1993). Since previous research has demonstrated associations between occupational L4/L5 reaction shear force exposures occupational low-back pain reporting (Norman et al. 1998), it was not deemed necessary to employ more sophisticated musculoskeletal modeling techniques (e.g., Arjmand et al. 2010; Granata and Marras 2000; Staudenmann et al. 2007) in this experiment.

To define the initiation ("START" in Figure 4.3) and termination ("END" in Figure 4.3) of each lifting exertion, signals from the lifting switch and various kinematic profiles (e.g., whole-body centre-of-mass trajectory, spatial and temporal angular kinematics of the lumbar spine, etc.) were used to derive an event-detection algorithm in Visual3D™. Lift initiation was defined as the frame preceding the descent towards grasping the handles of the lifting apparatus (i.e., the last upright standing frame before bending forward), and lift termination was defined as the frame at which the lifting apparatus was placed at the destination target based on the switch signal (Figure 4.3). A third event, within the "window" bounded by the initiation and termination end-points, signified the frame at which the peak L4/L5 compression force was imposed ("PEAK" in Figure 4.3). To verify that events were defined as desired, model animations and the abovementioned kinematic and kinetic time-series data were visually inspected in Visual3D™ by a research assistant.

Peak L4/L5 bone-on-bone compression and reaction shear forces were extracted from the time-series data associated with each lifting exertion (between lift START and lift END). Total motion of each "joint" was calculated as the difference between the maximum and minimum angle captured during a complete lifting exertion. Since intervertebral joint failure load tolerance is reduced in non-neutral postures (Gallagher et al. 2005; Gunning et al. 2001; Howarth and Callaghan 2011), the absolute value of the lumbar spine angle at PEAK frame was also recorded. The orientation of the lumbar spine in a relaxed upright standing trial was defined as zero degrees about the flexion/extension, lateral bend, and axial twist axes. Mean values were calculated from three repetitions of each lifting task; these values constituted the dependent variables in the statistical analyses.

Statistical Analyses

Using the general linear model procedure in SAS system software (Windows Version 9.1.3 with Service Pack 4, SAS Institute Inc., Cary, NC, United States), within-subject comparisons were performed to examine the effects of condition (no-brace/brace), load (light/heavy), and the interaction of condition×load on lifting kinematics and kinetics. Least-square means were computed when condition×load effects were statistically significant; adjustments for multiple comparisons were performed using the Tukey method. In all statistical tests, the null hypothesis was rejected if the *p*-value was less than 0.05.

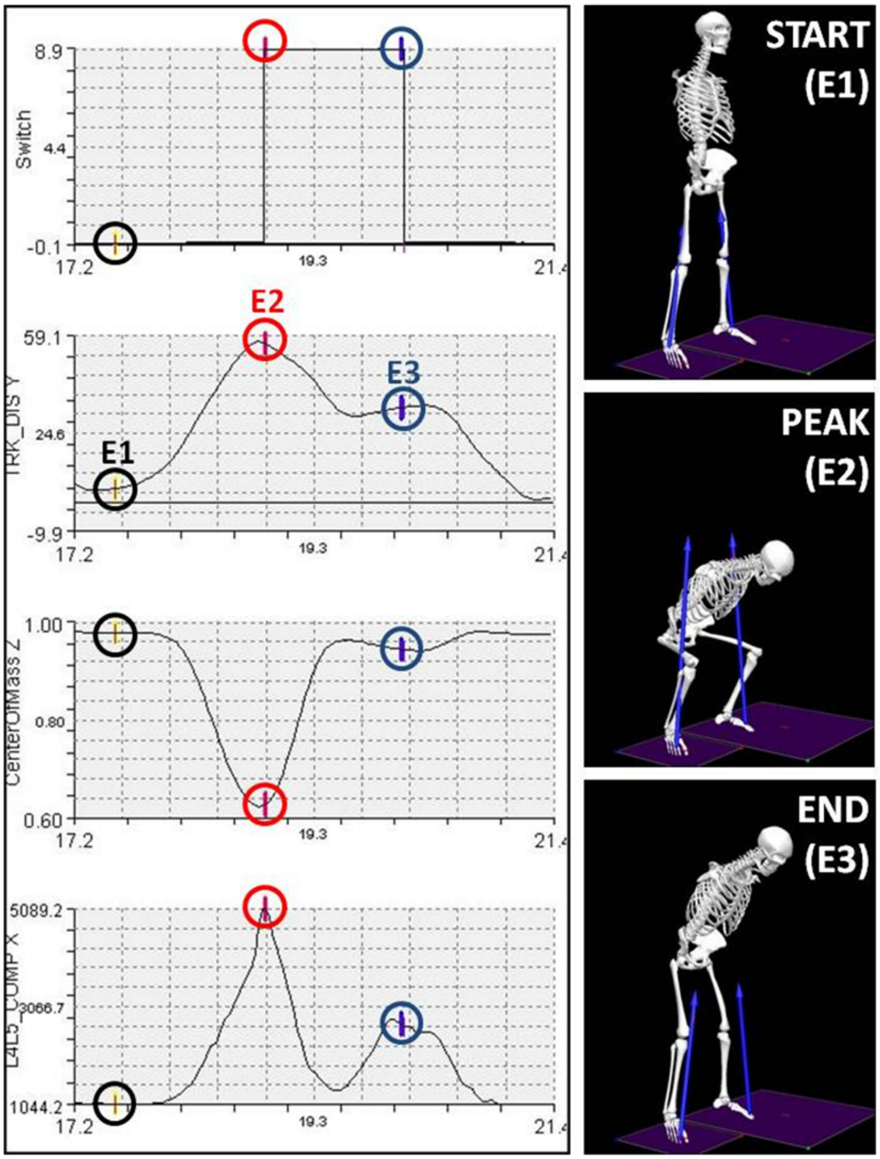

Figure 4.3. Events corresponding to the initiation (START = E1) and termination (END = E3) of each lifting exertion were defined, and the time at which the peak L4/L5 joint compressive force was imposed (PEAK = E2) was also registered.

4.3. Results

Despite the potential for complex interactions between the effects of ankle immobilization and mass lifted, there were only 10 out of a possible 216 cases (9 tasks X 24 dependent variables/task) where statistically significant condition×load effects were detected in the kinematic variables of interest. There were no statistically significant condition×load effects detected in any of the peak low-back loading variables. This indicated that the effects of ankle immobilization were effectively consistent between the 3.7 and 12.7 kg lifting tasks. Given the objective of this study, the impact of ankle immobilization is emphasized below rather than independent effects associated with mass lifted.

Group Kinematic Response

Compensatory movement strategies varied somewhat depending on the external task demands (i.e., lift origins and destinations), however a number of common kinematic responses were observed when comparisons were made between the conditions (no-brace vs. brace). Compared to the no-brace condition, subjects utilized less ankle (left = 3.4°; right = 20.8°), less knee (left = 15.5°; right = 19.1°), greater ipsilateral hip (2.0°), and greater lumbar spine (2.0°) sagittal plane motion when lifting (Figure 4.4). As summarized in Tables 4.2 to 4.5, statistically significant differences in sagittal plane motion were generally of greater magnitude and more consistently observed on the right side of the body and when lifting from Positions 2 and 3. However, statistically significant differences ($p < 0.05$) were also detected in a number of frontal and transverse plane joint motion variables and across all lifting tasks (Tables 4.3 to 4.5).

Group Low-Back Loading Response

When lifting from Positions 2 and 3 (i.e., from directly in-front or from the right side of the subject), peak L4/L5 A/P reaction shear forces were on average 23% (74 N) greater ($p \leq 0.0387$) when wearing the right ankle brace (compared to the no-brace condition) (Figure 4.5). A similar response resulted when lifting from Position 1 to 4 ($p = 0.0483$), but between-condition (no-brace vs. brace) differences in peak L4/L5 A/P reaction shear forces were not detected when lifting from Position 1 to 5 ($p = 0.2450$) or from Position 1 to 6 ($p = 0.1057$). Irrespective of the external task demands, there were no statistically significant differences in peak L4/L5 compressive ($0.0816 \le p \le 0.9503$) or M/L shear ($0.0698 \le p \le 0.9538$) forces between the brace and no-brace conditions (Figures 4.6 and 4.7).

When wearing the ankle brace, there was a general tendency for the lumbar spine to be more deviated when the peak L4/L5 compression force was imposed. However, between-condition (i.e., no-brace vs. brace) differences were of relatively small magnitudes and were not statistically significant in all tasks or about all anatomical axes (Table 4.6).

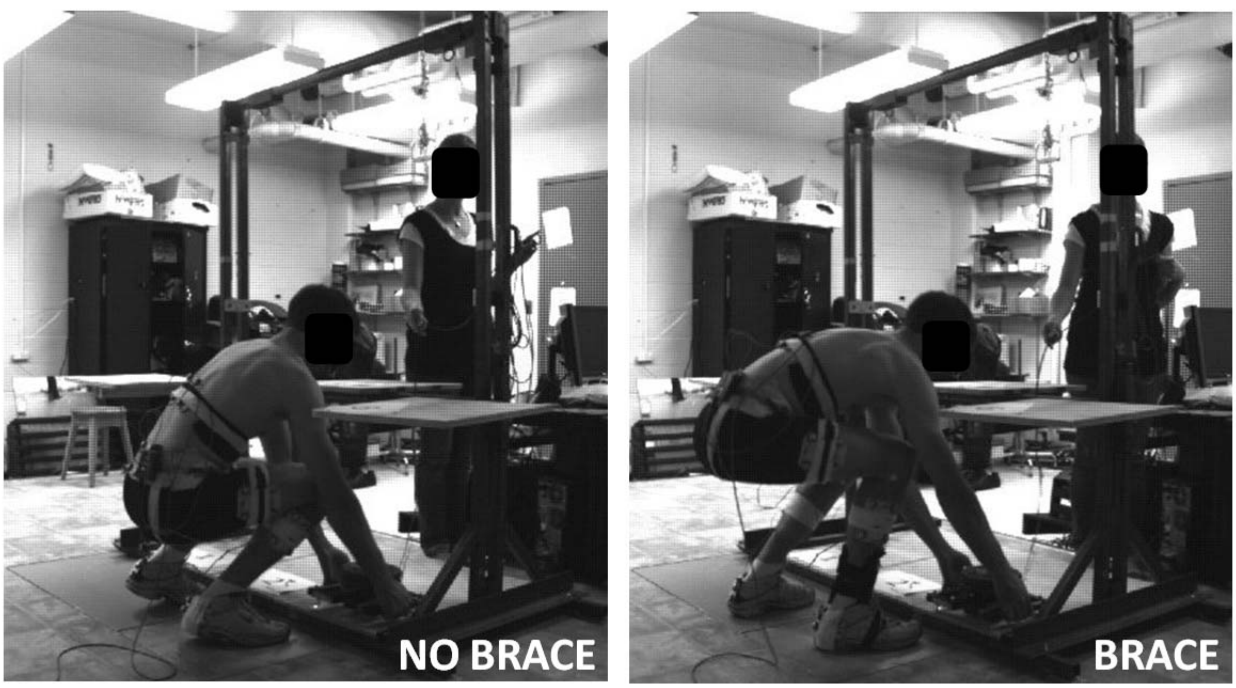

Figure 4.4. Depiction of the “group” response to unilateral ankle immobilization. When wearing the right ankle brace, subjects exhibited less ankle, less knee, greater ipsilateral hip, and greater lumbar spine motion than they did when lifting without the brace.

Table 4.2. Total (peak-to-peak) angular motion about the lumbar spine during a complete lifting exertion. Data represent the mean (SEM) values calculated across all subjects (N = 10).

Task	Frontal Motion (degrees)			Sagittal Motion (degrees)			Transverse Motion (degrees)		
	No Brace	**Brace**	***p*-value**	**No Brace**	**Brace**	***p*-value**	**No Brace**	**Brace**	***p*-value**
1to4	10.7 (0.7)	11.1 (0.7)	0.2905	56.5 (2.7)	58.8 (2.4)	0.0500	5.3 (0.3)	6.1 (0.3)	0.0043
1to5	6.8 (0.4)	7.3 (0.3)	0.4080	58.0 (2.6)	59.2 (2.5)	0.2502	4.8 (0.3)	5.9 (0.4)	0.0133
1to6	14.1 (0.9)	15.2 (0.9)	0.3242	58.0 (2.4)	59.0 (2.5)	0.3491	6.6 (0.5)	6.8 (0.4)	0.6116
2to4	11.3 (0.9)	10.4 (0.7)	0.2248	54.0 (3.0)	55.5 (2.6)	0.3265	4.7 (0.3)	4.8 (0.4)	0.9037
2to5	4.3 (0.6)	4.8 (0.4)	0.2716	52.8 (3.0)	55.4 (2.6)	0.0630	3.6 (0.3)	3.5 (0.3)	0.6980
2to6	10.9 (0.7)	11.7 (0.7)	0.0874	54.1 (3.0)	55.6 (2.6)	0.2315	4.5 (0.3)	4.8 (0.4)	0.0523
3to4	14.5 (1.2)	15.3 (1.1)	0.2946	56.2 (2.8)	59.3 (2.6)	0.0006	5.7 (0.4)	6.3 (0.5)	0.3175
3to5	7.2 (0.8)	6.9 (0.9)	0.6724	56.1 (2.6)	57.7 (2.6)	0.0307	4.7 (0.3)	5.1 (0.4)	0.2473
3to6	9.5 (0.7)	10.8 (0.9)	0.1015	56.4 (3.0)	59.6 (2.1)	0.0628	5.6 (0.6)	6.0 (0.6)	0.3792

Table 4.3. Total (peak-to-peak) angular motion about the ankle during one complete lifting exertion. Data represent the mean (SEM) values calculated across all subjects (N = 10).

Task	Side	Frontal Motion (degrees)			Sagittal Motion (degrees)			Transverse Motion (degrees)		
		No Brace	**Brace**	***p*-value**	**No Brace**	**Brace**	***p*-value**	**No Brace**	**Brace**	***p*-value**
1to4	Left	9.1 (0.3)	9.5 (0.4)	0.2941	28.4 (1.7)	24.9 (2.0)	0.0112	11.4 (0.8)	10.9 (0.8)	0.6072
	Right	6.9 (0.4)	4.4 (0.5)	0.0003	32.9 (1.5)	11.1 (0.8)	0.0001	12.1 (0.7)	8.7 (0.7)	0.0228
1to5	Left	9.3 (0.5)	10.1 (0.7)	0.1498	30.6 (1.5)	28.5 (2.2)	0.1199	9.0 (0.6)	9.0 (0.7)	0.7153
	Right	6.0 (0.6)	4.4 (0.8)	0.0249	27.9 (1.9)	9.1 (1.0)	<0.0001	10.2 (0.6)	8.5 (0.9)	0.1902
1to6	Left	12.4 (0.8)	12.3 (0.8)	0.7294	33.7 (2.3)	29.1 (2.1)	0.0196	9.9 (0.8)	10.9 (0.9)	0.1453
	Right	9.3 (0.7)	6.9 (0.7)	0.0054	25.7 (2.0)	10.6 (1.1)	0.0001	12.4 (0.9)	8.2 (0.9)	0.0018
2to4	Left	8.3 (0.4)	7.5 (0.6)	0.1193	31.7 (1.8)	27.9 (1.9)	0.0037	13.4 (1.1)	13.2 (1.0)	0.8422
	Right	9.8 (0.6)	5.1 (0.4)	<0.0001	35.7 (1.9)	12.6 (0.8)	<0.0001	11.5 (0.9)	6.8 (0.6)	0.0009
2to5	Left	6.7 (0.7)	5.5 (0.6)	0.0372	33.8 (1.5)	30.0 (2.1)	0.0945	9.3 (0.7)	10.1 (0.7)	0.2620
	Right	7.1 (0.5)	3.4 (0.3)	<0.0001	35.8 (1.3)	12.0 (1.1)	<0.0001	10.9 (0.7)	6.4 (0.9)	0.0273
2to6	Left	9.1 (0.6)	8.6 (0.9)	0.4326	32.5 (1.7)	27.3 (1.9)	0.0072	9.7 (0.7)	9.8 (0.6)	0.8353
	Right	8.0 (0.4)	5.3 (0.4)	0.0006	32.8 (1.6)	11.1 (1.3)	<0.0001	15.1 (1.2)	7.5 (0.8)	0.0014
3to4	Left	9.5 (0.8)	8.9 (0.7)	0.1862	29.3 (1.9)	27.7 (2.2)	0.4424	11.6 (1.2)	11.6 (1.3)	0.9930
	Right	11.7 (0.6)	7.5 (0.6)	<0.0001	35 (2.2)	13.6 (0.8)	<0.0001	12.2 (0.7)	7.7 (0.4)	0.0013
3to5	Left	6.7 (0.5)	6.9 (0.7)	0.6905	31.7 (1.7)	29.8 (2.0)	0.2160	9.4 (0.7)	10.3 (0.7)	0.0915
	Right	8.1 (0.5)	6.1 (0.6)	0.0090	35.2 (2.0)	12.7 (1.0)	<0.0001	11.5 (0.6)	6.5 (0.6)	0.0045
3to6	Left	6.9 (0.5)	7.0 (0.8)	0.9048	32 (1.8)	28.2 (2.2)	0.0194	10.0 (0.7)	10.0 (0.9)	0.9974
	Right	8.6 (0.4)	6.3 (0.6)	0.0001	30.8 (1.7)	11.5 (0.9)	<0.0001	13.5 (0.7)	6.2 (0.5)	<0.0001

Table 4.4. Total (peak-to-peak) angular motion about the knee during one complete lifting exertion. Data represent the mean (SEM) values calculated across all subjects (N = 10).

Task	Side	Frontal Motion (degrees)			Sagittal Motion (degrees)			Transverse Motion (degrees)		
		No Brace	Brace	*p*-value	No Brace	Brace	*p*-value	No Brace	Brace	*p*-value
1to4	Left	13.9 (2.0)	13.4 (2.0)	0.0773	100.1 (6.8)	83.4 (7.1)	0.0078	17.5 (3.5)	14.0 (2.6)	0.1371
	Right	10.5 (1.3)	9.5 (1.4)	0.0414	102.2 (6.9)	80.9 (6.6)	0.0064	10.9 (1.3)	8.3 (1.4)	0.0655
1to5	Left	13.9 (2.1)	13.7 (2.1)	0.0780	95.7 (6.6)	87.4 (7.7)	0.0550	16.3 (3.5)	14.7 (2.6)	0.2344
	Right	10.0 (1.3)	10.3 (1.8)	0.6673	92.5 (7.3)	81.4 (7.9)	0.0335	9.6 (1.2)	8.4 (1.4)	0.2065
1to6	Left	15.3 (2.1)	14.4 (2.2)	0.0183	97.1 (7.2)	87.8 (7.4)	0.0354	17.8 (3.5)	14.9 (2.9)	0.1121
	Right	10.5 (1.3)	10.3 (1.7)	0.8864	87.9 (7.4)	78.5 (7.5)	0.0697	9.6 (1.2)	7.7 (1.0)	0.0201
2to4	Left	14.8 (2.0)	13.2 (2.0)	0.0007	104.7 (4.9)	87.8 (6.2)	0.0015	18 (3.4)	13.4 (2.6)	0.0541
	Right	10.4 (1.4)	9.7 (1.5)	0.1230	104.6 (4.5)	84.3 (5.9)	0.0004	11.2 (1.1)	9.3 (1.4)	0.0707
2to5	Left	14.7 (1.9)	13.3 (2.0)	0.0032	109.2 (4.5)	89.1 (6.4)	0.0016	19.5 (3.2)	13.8 (2.8)	0.0136
	Right	10.5 (1.5)	9.5 (1.6)	0.0471	107.4 (4.8)	83.5 (6.0)	0.0005	12.1 (1.5)	9.2 (1.4)	0.0604
2to6	Left	15.2 (2.1)	14.1 (2.2)	0.0154	109.5 (5.2)	91.8 (6.5)	0.0017	20.0 (3.6)	14.0 (2.8)	0.0329
	Right	10.7 (1.3)	10.3 (1.6)	0.5743	103.4 (4.9)	84.5 (6.6)	0.0041	11.7 (1.2)	10.0 (1.4)	0.1200
3to4	Left	14.2 (1.8)	12.8 (1.8)	0.0530	99.7 (6.8)	82.7 (6.8)	0.0138	16.9 (3.3)	11.8 (2.3)	0.0841
	Right	9.7 (1.2)	10.0 (1.5)	0.6105	100.9 (6.4)	79.7 (6.0)	0.0019	9.9 (1.2)	9.4 (1.6)	0.6970
3to5	Left	14.0 (2.0)	13.1 (1.9)	0.0730	104.9 (6.0)	91.6 (7.3)	0.0471	18.2 (3.7)	14.0 (2.7)	0.0980
	Right	9.9 (1.2)	9.7 (1.4)	0.6453	103.1 (5.9)	83.0 (6.9)	0.0084	10.8 (1.2)	9.4 (1.5)	0.2983
3to6	Left	14.6 (2.1)	13.2 (2.1)	0.0277	111.3 (6.2)	91.3 (6.2)	0.0026	18.9 (3.7)	13.4 (2.8)	0.0132
	Right	9.9 (1.3)	9.1 (1.4)	0.0964	102.3 (5.9)	76.9 (5.3)	0.0015	11.0 (1.3)	8.5 (1.4)	0.1665

Table 4.5. Total (peak-to-peak) angular motion about the hip during one complete lifting exertion. Data represent the mean (SEM) values calculated across all subjects (N = 10).

Task	Side	Frontal Motion (degrees)			Sagittal Motion (degrees)			Transverse Motion (degrees)		
		No Brace	**Brace**	***p*-value**	**No Brace**	**Brace**	***p*-value**	**No Brace**	**Brace**	***p*-value**
1to4	Left	14.0 (1.1)	13.7 (1.1)	0.8653	93.3 (2.3)	91.6 (2.4)	0.1136	29.5 (1.8)	27.8 (2.1)	0.2421
	Right	22.7 (1.4)	23.9 (1.4)	0.4412	93.3 (2.6)	94.2 (2.8)	0.4673	46.5 (2.4)	39.6 (2.5)	0.0301
1to5	Left	10.1 (0.7)	11.0 (1.0)	0.4446	93.0 (2.5)	91.7 (2.2)	0.1815	24.8 (2.5)	22.2 (3.0)	0.0284
	Right	23.4 (1.5)	23.9 (1.7)	0.8962	86.4 (2.8)	87.5 (2.3)	0.5108	21.1 (2.0)	18.2 (1.2)	0.1834
1to6	Left	13.0 (0.7)	12.3 (0.7)	0.4495	96.7 (2.3)	96.4 (2.4)	0.7655	50.1 (3.6)	48 (3.5)	0.2107
	Right	28.2 (1.4)	26.9 (1.8)	0.4263	85.9 (2.7)	86.9 (2.7)	0.5479	35.8 (1.1)	37.8 (1.2)	0.2547
2to4	Left	18.3 (2.0)	18.5 (1.7)	0.9113	91.1 (2.7)	90.3 (2.1)	0.5287	31.3 (2.0)	30.2 (2.0)	0.1738
	Right	14.3 (1.0)	14.8 (1.4)	0.7546	94.5 (2.5)	97.7 (2.0)	0.0345	43.7 (2.0)	41.6 (1.9)	0.3335
2to5	Left	15.6 (1.3)	15.2 (1.3)	0.6881	91.6 (2.4)	91.4 (2.2)	0.8928	24.2 (3.0)	20.8 (3.3)	0.0757
	Right	15.5 (1.6)	13.7 (1.6)	0.3658	90.3 (2.1)	93.7 (2.3)	0.0412	20.2 (1.9)	16.6 (0.9)	0.1181
2to6	Left	16.7 (1.1)	16.2 (1.2)	0.6247	94.2 (2.6)	93.3 (2.2)	0.4992	49.1 (3.9)	45.4 (3.7)	0.0134
	Right	16.6 (1.5)	17.9 (2.2)	0.5231	90.4 (2.2)	93.2 (2.1)	0.0112	33.8 (1.9)	32.4 (0.7)	0.5616
3to4	Left	28.2 (1.8)	27.8 (1.6)	0.6612	86.9 (2.9)	85.4 (2.7)	0.3829	38.5 (1.8)	35.6 (1.8)	0.0176
	Right	11.8 (0.8)	13.1 (1.2)	0.2497	98.7 (2.1)	99.8 (2.3)	0.1579	47.3 (2.3)	44.6 (1.6)	0.3075
3to5	Left	24.6 (1.1)	24.9 (1.3)	0.6426	87.7 (2.7)	88.2 (2.2)	0.7093	26.1 (3.3)	22.5 (3.4)	0.0728
	Right	9.4 (0.7)	11.4 (1.2)	0.0872	93.1 (2.1)	95.5 (2.0)	0.0060	22.6 (2.1)	20.8 (1.2)	0.2997
3to6	Left	24.3 (1.1)	25.5 (1.3)	0.4291	91.7 (2.5)	93.2 (2.6)	0.3242	49.7 (4.3)	42.7 (4.1)	0.0008
	Right	11.7 (0.5)	12.8 (1.2)	0.3267	92.8 (1.8)	95.1 (2.0)	0.0492	29.2 (1.3)	29.5 (1.4)	0.8972

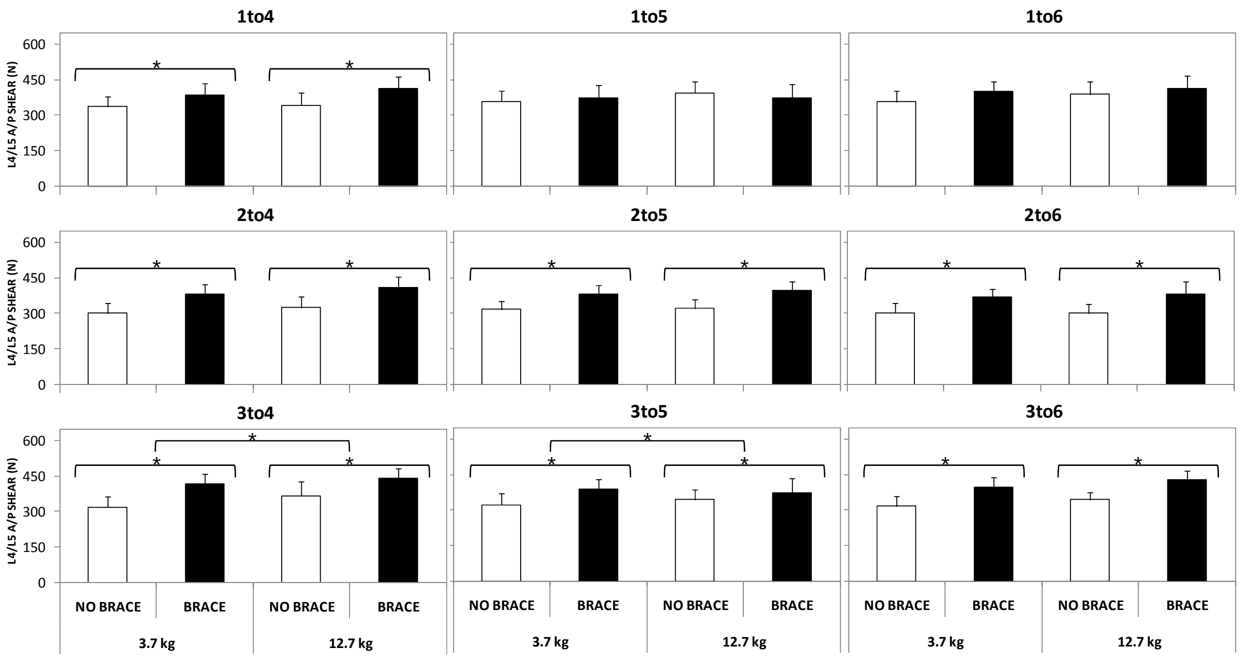

Figure 4.5. Group summary of the peak low-back A/P shear forces calculated during lifting. Results from all lifting tasks (all possible origin-to-destination combinations) and all participants (N = 10) are included. Data represent the mean values across all participants; error bars represent the standard error of the mean. * indicates that differences were statistically significant ($p < 0.05$).

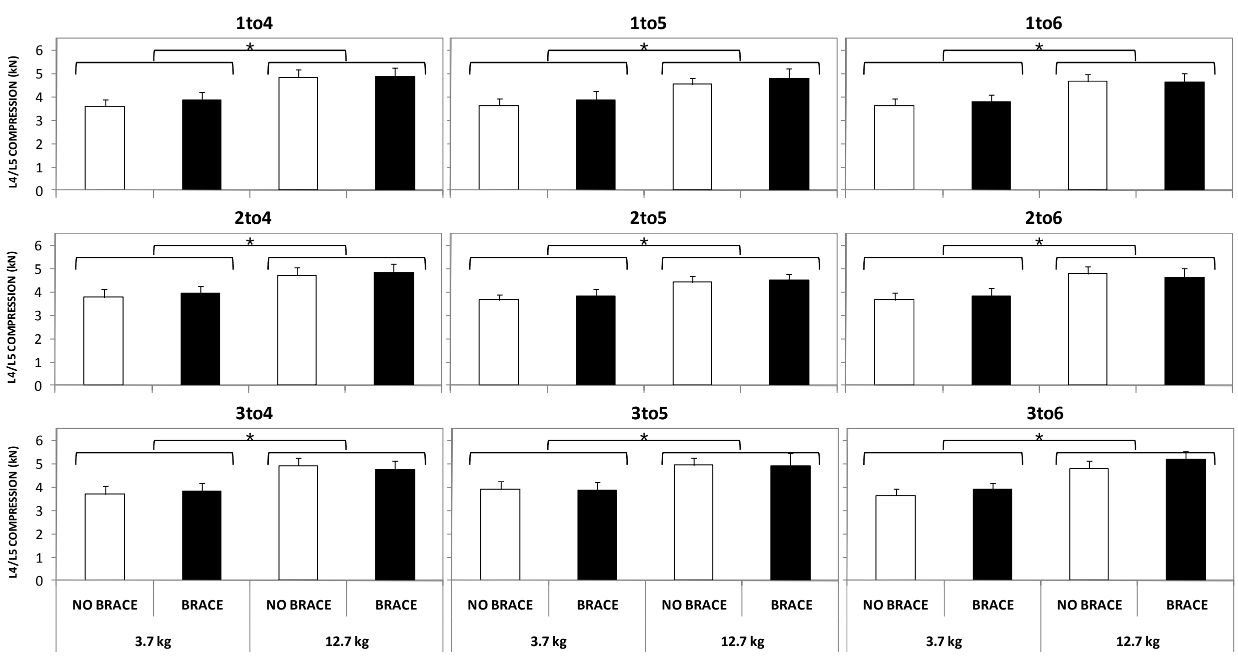

Figure 4.6. Group summary of the peak low-back compressive forces calculated during lifting. Results from all lifting tasks (all possible origin-to-destination combinations) and all participants (N = 10) are included. Data represent the mean values across all participants; error bars represent the standard error of the mean. * indicates that differences were statistically significant ($p < 0.05$).

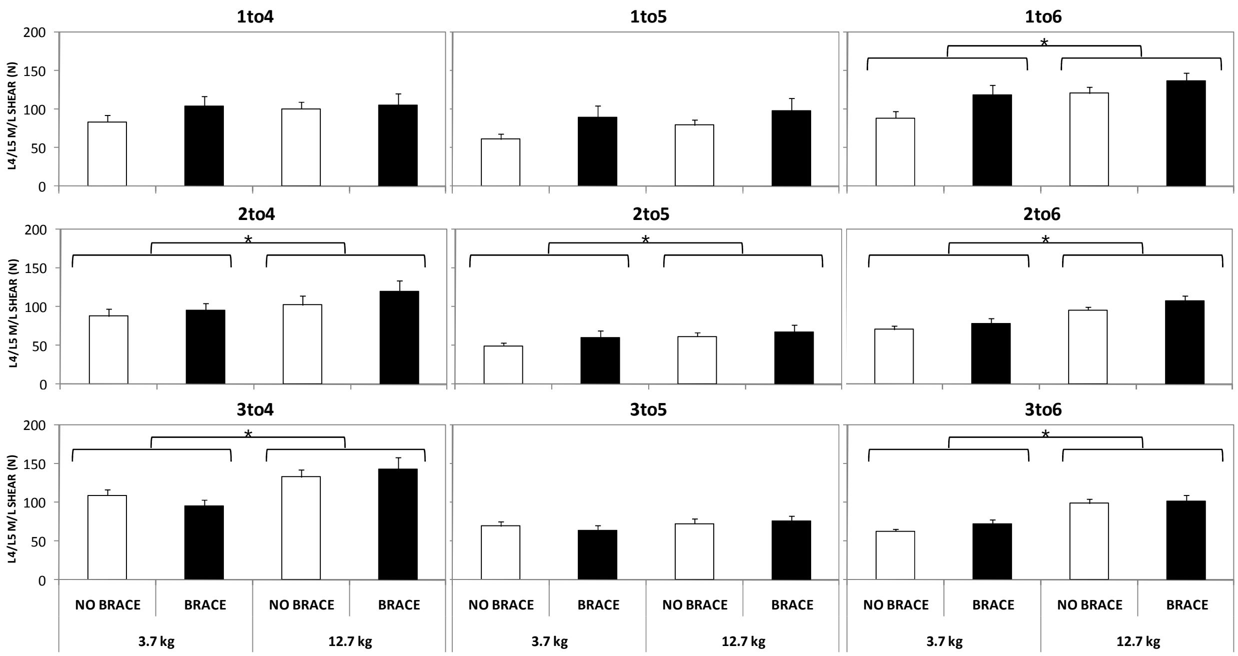

Figure 4.7. Group summary of the peak low-back M/L shear forces calculated during lifting. Results from all lifting tasks (all possible origin-to-destination combinations) and all participants (N = 10) are included. Data represent the mean values across all participants; error bars represent the standard error of the mean. * indicates that differences were statistically significant ($p < 0.05$).

Case Study

Although some inter-individual differences in compensatory movement strategies were observed when the right ankle was immobilized, subjects usually exhibited rather subtle variations of the "group" kinematic response described above. However, there were several instances when individuals responded somewhat differently to the imposed ankle motion constraint. On occasion (usually when lifting from the right side, Position 3), a subject would utilize greater bilateral knee and hip sagittal plane motion when wearing the ankle brace. This response was also characterized by more right "heel-lift" than was observed in the no-brace condition, an observation that was inconsistent with the "group" kinematic response (Figure 4.4). Interestingly, the low-back compressive and A/P shear loading responses associated with these different compensatory movement strategies also varied from the "group" response; the way in which an individual compensated influenced the L4/L5 joint loading pattern when lifting. In Figure 4.8, the low-back loading responses of two similarly-sized subjects who best represented the "group" (Strategy 1 in Figure4.8) and "heel-lift" (Strategy 2 in Figure 4.8) kinematic strategies are compared. In the examples provided, it can be seen that the "group" kinematic response to right ankle immobilization was associated with greater L4/L5 A/P reaction shear forces and no biomechanically meaningful changes in L4/L5 joint compression, whereas the opposite loading pattern was associated with the "heel-lift" strategy exampled.

Table 4.6. Angular deviation of the lumbar spine when the peak L4/L5 joint compression force was applied. Absolute values of the lateral bend (Frontal Plane), flexion (Sagittal Plane), and axial twist (Transverse Plane) angles at PEAK frame were recorded. Data represent the mean (SEM) values calculated across all subjects (N = 10).

Task	Frontal Plane (degrees)			Sagittal Plane (degrees)			Transverse Plane (degrees)		
	No Brace	Brace	*p*-value	No Brace	Brace	*p*-value	No Brace	Brace	*p*-value
1to4	2.8 (0.5)	3.8 (0.5)	0.0036	55.9 (3.9)	58.2 (3.4)	0.0511	2.8 (0.6)	2.9 (0.7)	0.6365
1to5	2.2 (0.5)	4.4 (0.8)	0.0154	57.0 (4.0)	58.6 (3.7)	0.3616	2.3 (0.5)	3.4 (0.5)	0.0720
1to6	2.8 (0.4)	5.3 (1.0)	0.0194	57.4 (3.5)	58.6 (3.5)	0.2294	3.1 (0.6)	3.8 (0.7)	0.2328
2to4	2.7 (0.6)	2.7 (0.8)	0.8810	52.3 (4.4)	54.9 (3.8)	0.2016	2.0 (0.5)	1.5 (0.6)	0.2105
2to5	2.5 (0.4)	2.9 (0.7)	0.4298	51.9 (4.6)	54.6 (3.8)	0.0461	1.7 (0.5)	1.8 (0.6)	0.7516
2to6	2.2 (0.4)	3.2 (1.0)	0.2940	53.4 (4.5)	55.0 (3.8)	0.2127	1.8 (0.4)	2.0 (0.6)	0.5382
3to4	5.9 (1.3)	5.6 (1.3)	0.7795	55.7 (4.0)	57.7 (3.6)	0.0664	1.8 (0.3)	2.8 (0.5)	0.0251
3to5	5.6 (1.2)	4.8 (1.4)	0.4978	55.8 (3.7)	56.7 (3.7)	0.2762	2.1 (0.4)	2.8 (0.5)	0.0424
3to6	5.3 (1.3)	4.4 (1.2)	0.3806	55.5 (4.1)	58.4 (3.0)	0.1078	1.9 (0.5)	2.3 (0.5)	0.5143

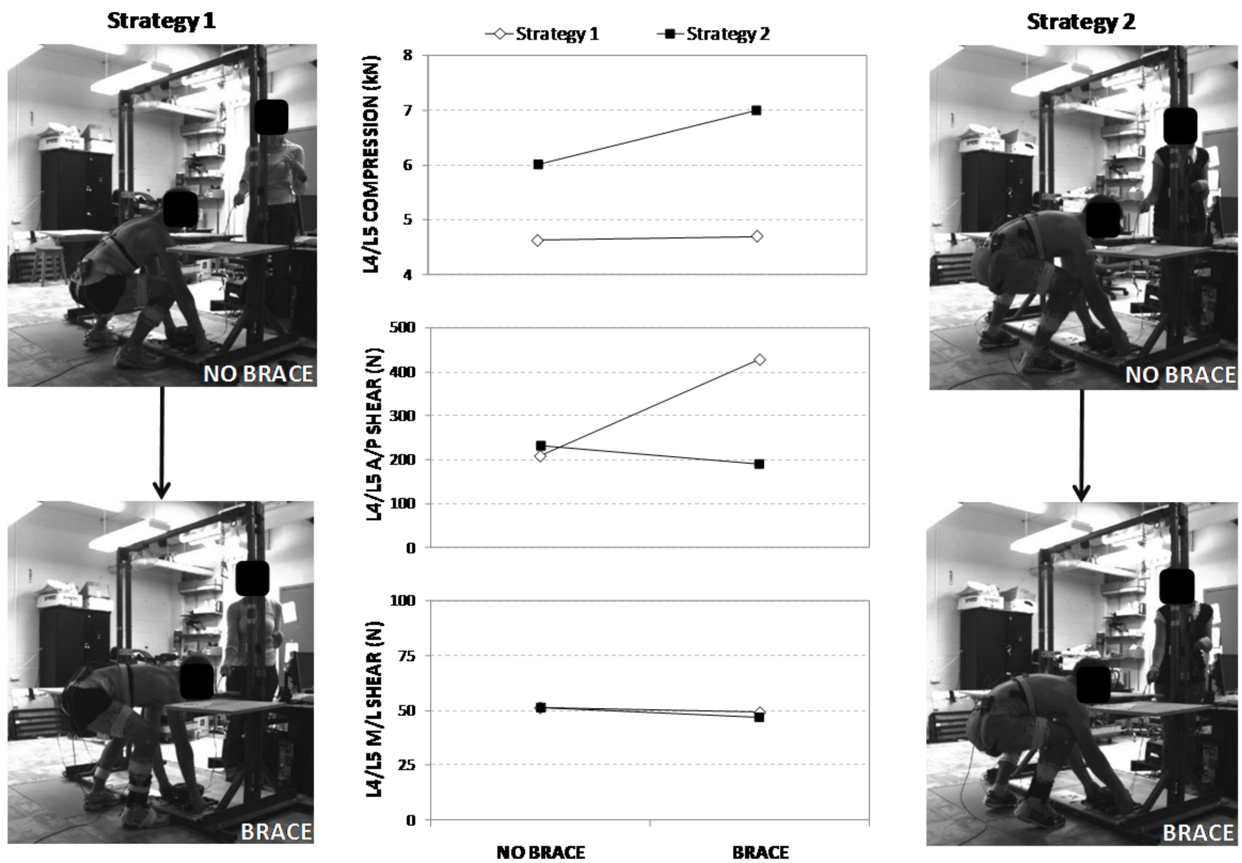

Figure 4.8. Low-back loading patterns were dependent on the compensatory movement strategy employed. Strategy 1 represents the "group" response, whereas Strategy 2 represents the alternative "heel-lift" strategy.

4.4. Discussion

Tested in this study was the influence of unilateral ankle immobilization on the peak low-back loading response to lifting. As hypothesized, subjects adapted their preferred movement strategy to compensate for the kinematic constraint and the low-back loading pattern changed as a consequence. A case study was presented to demonstrate that different compensatory movement strategies resulted in different low-back loading patterns. Marras and colleagues previously demonstrated that individual factors such as gender (Marras et al. 2002), personality type (Marras et al. 2000), work experience (Marras et al. 2006), and history of low-back troubles (Marras et al. 2001a) can influence the low-back loading response of workers; results of this study highlight an additional personal characteristic (distal lower extremity joint dysfunction) that could alter low-back loading patterns at work. Effectively, data reported in this investigation support the notion that occupational injury prevention approaches could be enhanced if consideration is given to the kinematic chain theory of human movement.

Some surgical procedures (e.g., tibiotalocalcaneal fusion due to failed ankle arthrodesis) can lead to lasting deficits in ankle mobility (Galey and Sferra 2002). Although such surgeries may be rare and it is unlikely that most industrial workers would be expected or permitted to perform occupational duties such as manual lifting following such procedures, observations made during a parallel research project motivated the author to examine the influence of unilateral ankle immobilization on low-back loading during lifting. Specifically, while viewing firefighters performing various manual handling tasks, an individual with severely restricted right ankle mobility (due to previous surgery) was observed to kinematically compensate in ways hypothesized to alter his potential for sustaining a work-related low-back injury. When the firefighter grasped the handles of a crate he was about to lift from the ground, he exhibited greater knee, hip, and lumbar spine flexion than did his coworkers while his right heel rose from the floor (Figure 4.9). In the current study, a custom ankle brace was applied to replicate the abovementioned functional limitation in student volunteers who were free of lower extremity pain or dysfunction. When the brace was applied, subjects tended to compensate in a manner that was different than the firefighter (Strategy 1 in Figure 4.8). There were, however, instances where the compensatory movement patterns of study subjects resembled what was observed in the parallel study (Strategy 2 in Figure 4.8). Albeit interesting that the "group" kinematic response differed from that of the firefighter, it is not surprising that there were inter- and intra-individual variations in the adaptive responses to ankle immobilization. Between-subjects differences in flexibility, strength, and balance capabilities are just a few of the possible factors that could have influenced individual responses, and within-subject variations were likely related to the fact that the exposure was novel and short-lived (i.e., it cannot be assumed that subjects would exhibit stable movement behaviour without opportunities to explore a wider range of possible movement solutions over multiple exposures). Nevertheless, the finding most relevant to the study objective was that the peak low-back loading response was *different* when compared between the brace and no-brace conditions. In the examples presented in Section 3.3, differences in peak L4/L5 A/P reaction shear (≈ 100% increase = 200N, Strategy 1 in Figure 4.8) and bone-on-bone compression (≈ 20% increase = 1000 N, Strategy 2 in Figure 4.8) forces resulted when the right ankle was immobilized. Changes of this magnitude are meaningful, especially if applied repeatedly (Norman et al. 1998) or if the lumbar spine is deviated from its neutral orientation (Gallagher et al. 2005; Gunning et al. 2001; Howarth and Callaghan 2011).

Figure 4.9. A firefighter with restricted right ankle mobility (due to previous surgery) lifting a crate from the floor.

The "group" response to right ankle immobilization did not result in different peak low-back compressive forces during lifting. However, limitations associated with the biomechanical modeling approach employed must be considered before accepting this finding at face value. A three-dimensional net joint L4/L5 moment of force was calculated using a "bottom-up" inverse dynamical linked-segment model of the lower body, and the orthogonal components of the L4/L5 moment were input into a polynomial (McGill et al. 1996b) to incorporate muscular contributions to the joint compressive force. Although developed using an anatomically-detailed EMG-driven musculoskeletal modeling approach (Cholewicki and McGill 1996), the polynomial model was clearly insensitive to potential inter- or intra-individual differences in trunk muscle activation patterns. Given that lower body and lumbar spine kinematic compensatory strategies were observed when the ankle was immobilized, it is possible that trunk muscle activities (and their contributions to low-back compression) varied in response to the experimental treatment but were undetected using the modeling approach employed. Furthermore, because low-back A/P reaction shear forces were affected by ankle immobilization, contributions from potential external shear-offsetting muscles (e.g., lumbar erector spinae) to the L4/L5 bone-on-bone force (Potvin et al. 1991a; Potvin et al. 1991b) could also have been different between the brace and no-brace conditions. Future work incorporating measures of torso muscle activation (e.g., surface electromyograms) in the estimation of low-back loading (c.f., Staudenmann et al. 2007) are warranted to address this limitation. Nevertheless, it is important to emphasize that when the ankle was immobilized, there was a tendency for the lumbar spine to be more deviated when peak L4/L5 compression force magnitudes were imposed. This suggests that even if low-back compressive loading demands remained unaffected by ankle immobilization, the capacity to withstand imposed demands could have been compromised (Gallagher et al. 2005; Gunning et al. 2001).

At first glance, results of this study may not appear relevant to problems most frequently encountered by occupational injury prevention researchers or practitioners. Rather, it could be argued that results would be of more interest to return-to-work and disability prevention specialists who must develop strategies to effectively and efficiently rehabilitate and integrate previously injured individuals back into their work. However, the reader is encouraged to consider that personal movement constraints (i.e., limited joint mobility or neuromuscular control deficits) can also result from intrinsic or extrinsic factors unrelated to previous injury. For instance, individuals may possess limited ankle mobility due to acquired or inherited anatomical/structural deformities or neurological disorders (Cowley et al. 2009; Hill 1995; Sobel et al. 1997). Conversely, uniforms or personal protective equipment worn by workers (e.g., stiff safety boots) could also impose ankle motion restrictions (Böhm and Hösl 2010; Cikajlo and Matjacić 2007). Results of this study demonstrate how kinematic constraints could promote potentially injurious loading patterns distant from the affected site. Thus, for occupational injury prevention researchers and practitioners, this study provides further justification to consider the influence of personal factors on musculoskeletal injury potential.

4.5. Conclusions

In this experiment, it was demonstrated that distal lower extremity joint dysfunction (i.e., unilateral ankle immobilization) could alter low-back injury potential when lifting at work. Results reaffirm the classic notion that for many applications – including the science and practice of ergonomics – modeling the body as a "kinematic chain" can yield important information not always apparent during more targeted investigations or interventions. In occupations where modification of work tasks, environments, or systems is not practical or possible (e.g., firefighting), the kinematic chain theory of human movement can be applied to identify personal factors that influence the way workers move and load their low-back tissues.

CHAPTER 5

FMS™ Scores and Occupational Low-Back Loading Demands – Whole-Body Movement Screening as an Ergonomic Tool?

CHAPTER 5

FMS™ Scores and Occupational Low-Back Loading Demands – Whole-Body Movement Screening as an Ergonomic Tool?

Summary

Background: Previous research suggests that a general whole-body movement screen could be used to identify personal movement qualities that promote potentially injurious low-back loading patterns at work. The purpose of this study was to examine if Functional Movement Screen™ (FMS) scores could be used to project the low-back loading response to lifting.

Methods: Fifteen men who scored greater than 14 on the FMS (high-scorers) and 15 size-matched low-scorers (FMS < 14) performed sagittally symmetric and asymmetric laboratory-based lifting tasks. A three-dimensional dynamic biomechanical model was used to calculate peak L4/L5 joint forces, and the angle of the lumbar spine was recorded when the peak compressive force was applied.

Results: Regardless of the lifting task performed, there were no differences in peak L4/L5 joint compression ($p \geq 0.4157$), anterior/posterior reaction shear ($p \geq 0.5645$), or medial/lateral reaction shear ($p \geq 0.2581$) forces imposed on the low-backs of high- and low-scorers. At the instant when peak compressive forces were applied, the orientation of the lumbar spine was not different between high- and low-scorers about the lateral bend ($p \geq 0.4215$), axial twist ($p \geq 0.2734$), or flexion/extension ($p \geq 0.1354$) axes.

Conclusions: Using the previously established injury prediction threshold value of 14, the composite FMS score did not indicate who might overload their low-backs when lifting. Future attempts to modify or reinterpret FMS scoring are warranted given that several previous studies have revealed links between composite FMS scores and musculoskeletal complaints.

5.1. Introduction

In Chapter 4, it was demonstrated that distal lower extremity joint dysfunction (ankle immobilization) could influence low-back loading patterns and potential for injury when performing occupational lifting tasks. Despite being a logical (and thus predictable) response to eliminating biomechanical degrees-of-freedom in a closed kinematic chain, results of the experiment in the previous chapter provide empirical support for the notion that a general whole-body movement screen could be used to expose personal movement qualities that promote potentially injurious low-back loading patterns at work. When the body is viewed as a system (Davids et al. 2003), such qualities might represent inherent (endogenous) structural or functional attributes that effectively limit the number of movement strategies available to individuals and can thus influence how individuals consciously or subconsciously interact with their environment when engaged in physical activity (i.e., their movement behaviour).

The Functional Movement Screen™ (FMS) is a tool widely used to reveal a host of undesirable personal movement qualities (e.g., limited joint mobility, bilateral asymmetries, postural control deficits, and pain-producing movement patterns) (Cook et al. 2006a; Cook et al. 2006b; Cook et al. 2010). Using standardized verbal instructions, testing apparatus, and grading criteria, the ability to move freely, symmetrically, and without pain is appraised via visual inspection of whole-body movement strategies during FMS task execution. Patterns of coordination and control that differ overtly from those deemed normal or desirable are purported to indicate the presence of undesirable personal movement qualities that could, if left unchecked, promote potentially injurious movement patterns in life. Indeed, results of the experiment presented in Chapter 4 lend support for the notion that atypical or undesirable personal movement qualities could influence injury potential, as unilateral ankle immobilization altered the kinematics and kinetics of lifting.

Although relationships between FMS scores and musculoskeletal injury reporting are not always established (e.g., Burton 2006; Hoover et al. 2008; Krackow 2001; Sorenson 2009), results of several studies conducted by Kiesel and colleagues (Kiesel et al. 2006; Kiesel et al. 2007; Kiesel et al. 2009) suggest that the FMS could be useful as part of a comprehensive injury prevention program, especially given that a standardized exercise-based intervention exists to improve FMS scores (Kiesel et al. 2011). Kiesel et al. (2007) found that members of a professional American football roster who attained a composite FMS score of 14 or less were 11 times more likely to miss at least 3 weeks of competition due to injuries than were teammates who attained composite FMS scores greater than 14. Also noteworthy was that the probability of missing competition (due to injury) for players scoring 14 or less increased from 15 to 51% when compared to athletes with scores above 14. Kiesel et al. (2006) earlier reported that professional American football players who attained a composite FMS score of 14 or less were 11 times more likely to be injured during pre-season training than were teammates who attained composite FMS scores greater than 14. In a more recent study conducted during a 16-week firefighter training academy, Kiesel et al. (2009) found that the probability of missing training time due to a musculoskeletal complaint increased from 27 to 56% if an individual entered the academy with a composite FMS score of less than 14. Taken together with the outcomes of the experiment in the previous chapter, results of the studies reviewed above raised the following question: Can composite FMS scores be used to predict who might injure their low-backs when lifting at work?

In this study, the low-back loading response to lifting was compared between two groups of height- and weight matched subjects. Individuals who scored greater than 14 on the FMS, a previously established injury prediction threshold value, were allocated to the first group and size-matched counterparts who scored less than 14 were assigned to the second group. It was hypothesized that the peak low-back loading response to lifting would differ between the high- and low-scoring groups.

5.2. Methods

Subject Selection

In a larger study examining the influence of different exercise-based interventions on various fitness and low-back loading outcomes (Chapter 6), biomechanical data were collected from 60 male members of the Pensacola Fire Department (PFD, Pensacola, FL, United States) while they performed a battery of laboratory tasks at the Andrews-Paulos Research and Education Institute (APREI, Gulf Breeze, FL, United States). Due to equipment malfunction undetected at the time of collection, datasets from two subjects could not be used to address any questions posed in this thesis. All subjects signed an informed consent document that had been approved by the University of Waterloo's Office of Research Ethics, the Baptist Hospital Institutional Review Board, and the City of Pensacola.

Of the 58 subjects from which "usable" biomechanical data existed, 15 individuals achieved composite FMS scores greater than 14 and were assigned to a high-scoring group. From the remaining subjects (N = 43), 15 height- and weight-matched individuals with composite FMS scores less than 14 were assigned to a low-scoring group. A cut-off score of 14 was used to assign subjects to high- and low-scoring groups based on previous research which indicated that the probability of missing work- or sport-related training activities due to musculoskeletal complaints is up to 11 times greater in individuals who score less than 14 on the FMS (Kiesel et al. 2006; Kiesel et al. 2007; Kiesel et al. 2009). Physical characteristics of the 30 subjects included in this study and their composite FMS scores are listed in Table 5.1.

Although the intention was to use data from as many subjects as possible in this study, the number of high-scoring individuals and the desire to compare a balanced group of size-matched individuals limited the total sample size to 30. Size-matching was preferred to permit a direct between-group comparison of low-back loading magnitudes without the need to utilize scaling or normalization methods. Such methods would have been necessary to account for inter-subject differences in low-back loading that would have been due to differences in height and body mass.

Table 5.1. Age, height, mass, and composite FMS scores of study subjects.

Composite FMS Score > 14					Composite FMS Score < 14				
Subject	**Age (yrs)**	**Height (m)**	**Mass (kg)**	**FMS Score**	**Subject**	**Age (yrs)**	**Height (m)**	**Mass (kg)**	**FMS Score**
S10	25	1.81	83.7	16	S30	27	1.80	83.6	11
S34	22	1.83	85.6	16	S15	23	1.79	86.3	13
S47	46	1.74	100.2	16	S19	41	1.80	99.6	12
S68	21	1.74	76.4	18	S39	28	1.73	78.8	13
S73	25	1.79	94.5	15	S20	24	1.79	94.6	12
S18	36	1.77	83.2	16	S46	38	1.71	85.9	13
S63	39	1.66	76.1	16	S25	30	1.73	72.6	13
S65	44	1.72	76.1	16	S05	40	1.75	83.1	11
S55	34	1.85	91.6	15	S71	42	1.83	92.3	10
S58	45	1.74	78.5	15	S62	46	1.76	85.9	11
S22	28	1.88	113.7	15	S51	37	1.82	111.1	11
S13	38	1.78	89.9	18	S03	28	1.81	99.7	12
S69	25	1.87	119.6	18	S23	26	1.86	105.0	12
S17	47	1.80	91.3	16	S82	50	1.83	84.6	13
S02	28	1.81	77.7	16	S78	51	1.79	85.4	13
Mean (SD)	33.5 (2.4)	1.79 (0.02)	89.2 (2.5)	16.1 (0.3)	**Mean (SD)**	35.4 (2.4)	1.79 (0.01)	89.9 (2.7)	12.0 (0.3)

Experimental Protocol and Data Collection

In a separate session performed several weeks prior to the biomechanical data collection (described below), subjects were videotaped while performing FMS tasks. The videotaping took place at a PFD station in order to accommodate the variable schedules of the subjects and to permit the most practical way to complete the most movement screens in the least amount of time. (All movement screens were conducted over a 10-day period.) Two digital video cameras (Basler Inc., Exton, PA, United

States) were connected to a data acquisition computer running Vicon Nexus Motion Capture software (Version 1.5, Vicon, Oxford, United Kingdom); the software was used to synchronize, capture, and store the video files at a rate of 15 Hz. Video cameras were arranged at right angles to one another to capture FMS tasks from both frontal and sagittal plane perspectives. FMS tasks were performed 4 times in total – two repetitions of each FMS task were performed while subjects faced the frontal plane camera and two repetitions were performed while the subject faced away from the frontal plane camera. In this way, sagittal plane videos were recorded from both the left and right sides of subjects. Subjects were instructed to wear a t-shirt, shorts, and athletic shoes; however a number of subjects were required to perform the FMS tasks while wearing their PFD-issued station wear if they were on-shift. (This would allow them to respond promptly to an alarm.) The FMS was administered and graded exactly as described in Appendix II. As instructed by Cook et al. (2006a; 2006b; 2010), only the "best" (i.e., highest-graded repetition) was included in the composite score. All grading, performed by a single member of the research team who had over 8 years experience and training in FMS administration and interpretation, was performed post-collection via video observation. Anstee et al. (2003) reported that composite FMS scores are reliable when graded by a single trained observer on the basis of video recordings (intra-class correlation coefficient = 0.98).

In the biomechanics laboratory, subjects performed a variety of tasks ranging from general whole-body movements (e.g., squatting, lunging, pushing, pulling, etc.) to simulated job-specific duties (e.g., ceiling breach, forcible entry, equipment handling, etc.). However, only 24.7 kg and 9.3 kg symmetrical (sagittal) and asymmetrical lifts (Figure 5.1) were included in this investigation as a logical extension of the ankle immobilization study presented in the previous chapter. Laboratory-based lifting tasks were performed at a self-selected pace, and three repetitions of each lifting condition were recorded for analyses. Order of exposure was randomized between subjects and maximal recovery time was provided between exertions.

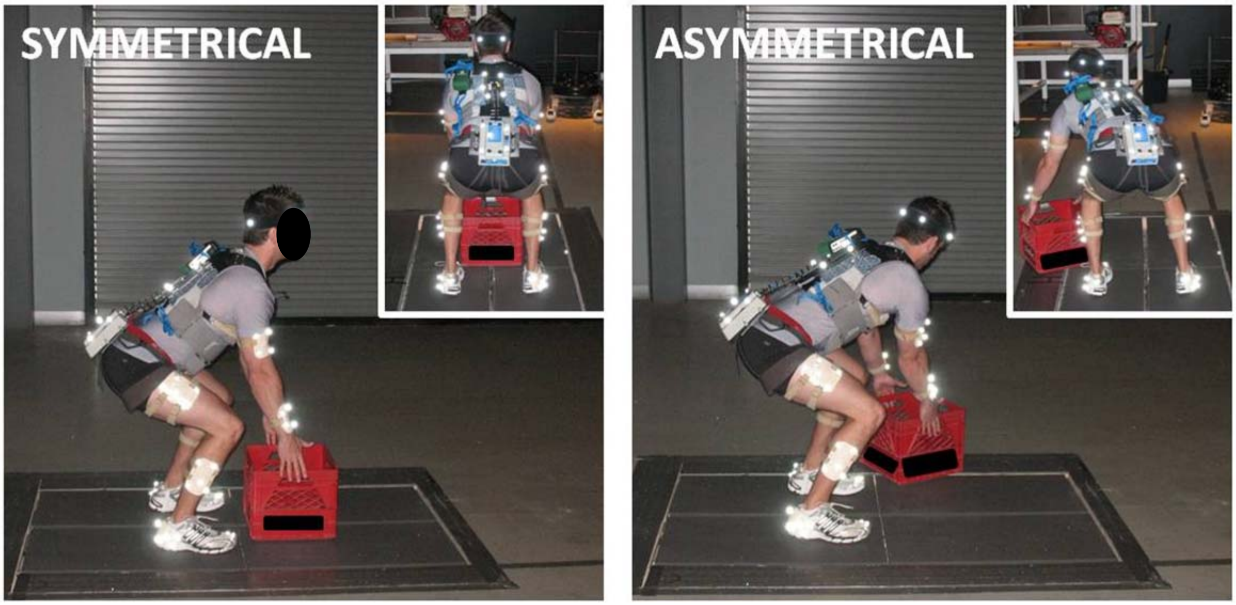

Figure 5.1. Two lifting tasks were performed by study subjects. The “symmetrical” task consisted of grasping and lifting crates (24.7 and 9.3 kg) that were located directly in-front of subjects, whereas the “asymmetrical” task was performed by grasping and lifting crates (24.7 and 9.3 kg) that were positioned at approximately 45 degrees to the left of the mid-sagittal plane.

To capture subject motion in the biomechanics laboratory, five reflective spherical markers (glued to custom-molded rigid bodies) were strapped to the feet, shanks, thighs, pelvis and trunk of subjects using double-sided tape and Velcro® straps. During a static (i.e., quiet standing) calibration trial collected prior to lifting, additional markers were taped to the skin overlying anatomical landmarks; these calibration markers were used to define the medial and lateral endpoints of modeled body segments (Figure 5.2). Anatomical landmarks coincided with recommendations made by Visual3D™ software developers (Version 4, C-Motion, Inc., Germantown, MD, United States). Marker position data were digitized at a frame frequency of 160 Hz using a 10-camera Vicon motion capture system (Vicon, Oxford, United Kingdom).

As shown in Figure 5.1, subjects performed the lifting tasks while standing on two force platforms (Bertec Corporation, Columbus, OH, United States). Using Vicon Nexus motion capture software (Version 1.5, Vicon, Oxford, United Kingdom), force platform signals were digitized at a rate of

2400 Hz and synchronized (spatially and temporally) with marker position data. Subjects lifted the crate from standardized positions, located adjacent to the force platforms.

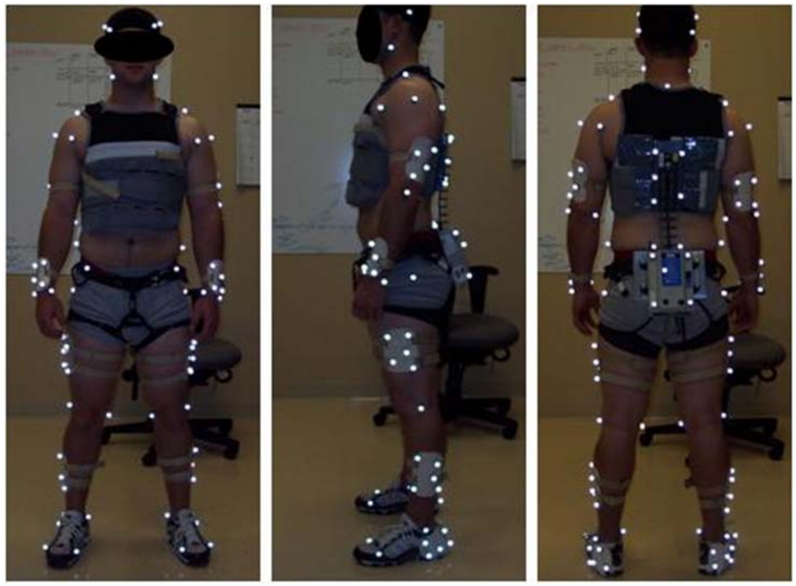

Figure 5.2. A whole-body marker set was applied to study subjects, but only data from the feet, shanks, thighs, pelvis, and trunk of study subjects were used to address questions posed in this thesis.

Data Processing and Analyses

Exactly as described in Chapter 3 using Visual3D™, a bottom-up inverse dynamical linked-segment modeling approach was used to calculate three-dimensional net L4/L5 joint reaction forces and moments of force. As described in Chapter 4, orthogonal components of the net L4/L5 moment were input into a polynomial model to compute dynamic L4/L5 joint ("bone-on-bone") compression forces (McGill et al. 1996b).

Event detection algorithms were generated using Visual3D™ by tracking the motion of the model centre-of-mass, trunk segment kinematics, and the L4/L5 joint compression force waveform. Events were created to signify the initiation ("START" in Figure 5.3) and termination ("END" in Figure 5.3) of each lift. Lift initiation was defined as the frame preceding the descent towards grasping the crate (i.e., the last upright standing frame before bending forward), and lift termination was defined as the frame at which the subject returned to upright standing whilst holding the crate. A third event, within the "window" bounded by the initiation and termination end-points, was defined to signify the frame at which the peak L4/L5 compression force was calculated ("PEAK" in Figure 5.3). To verify that events were defined as intended, model animations and the abovementioned kinematic and kinetic time-series data were visually inspected in Visual3D™.

From the time-series data associated with each lifting exertion (between lift START and lift END), peak L4/L5 joint compression and anterior/posterior (A/P) and medial/lateral (M/L) reaction shear forces were extracted. An additional variable of interest, the absolute value of the lumbar spine angle (angular displacement of the thorax with respect to the pelvis) at PEAK frame, was also extracted based on the knowledge that intervertebral joint failure load tolerance is reduced in non-neutral postures (Gallagher et al. 2005; Gunning et al. 2001; Howarth and Callaghan 2011). The orientation of the lumbar spine in a relaxed upright standing trial was defined as zero degrees about the flexion/extension, lateral bend, and axial twist axes. Mean values, computed using data acquired from three repetitions performed, constituted the dependent variables in the statistical analyses.

Statistical Analyses

Using the general linear model procedure in SAS system software (Windows Version 9.1.3 with Service Pack 4, SAS Institute Inc., Cary, NC, United States), between-subject (FMS group = high- vs. low-scorers) and within-subject (mass lifted = 24.7 vs. 9.3 kg) comparisons were made. (Each lifting position was analyzed separately.) If statistically significant group×load interaction effects were detected, a least-square means approach was used to make between-condition comparisons. Adjustments for multiple comparisons were performed using the Tukey method. In all statistical tests, the null hypothesis was rejected if the *p*-value was less than 0.05.

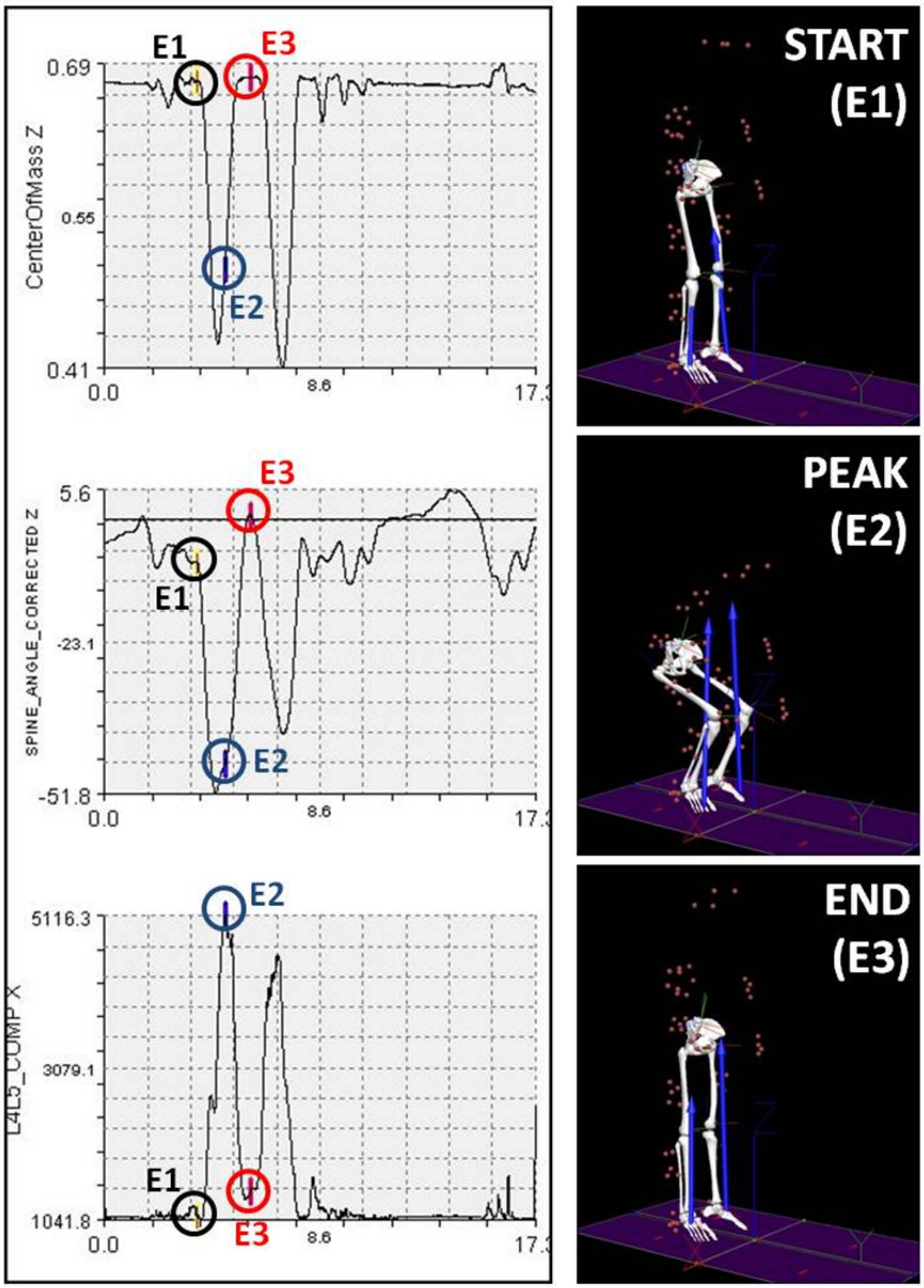

Figure 5.3. Events corresponding to the initiation (START = E1) and termination (END = E3) of each lifting exertion were defined, and the time at which the peak L4/L5 compressive force was imposed (PEAK = E2) was also registered.

5.3. Results

When performing the lifting tasks, peak loads imposed on the low-backs of individuals who scored less than 14 on the FMS (low-scorers) were not significantly different ($p \geq 0.2581$) from those applied to the low-backs of height- and weight-matched high-scorers (individuals who scored greater than 14 on the FMS) (Figure 5.4). Because there were no statistically significant differences in the mean age ($p = 0.6008$), height ($p = 0.9722$), or mass ($p = 0.8751$) between the high- and low-scoring FMS groups (achieved via experimental design), between-group comparisons in the unscaled magnitudes of peak low-back loading values were deemed permissible.

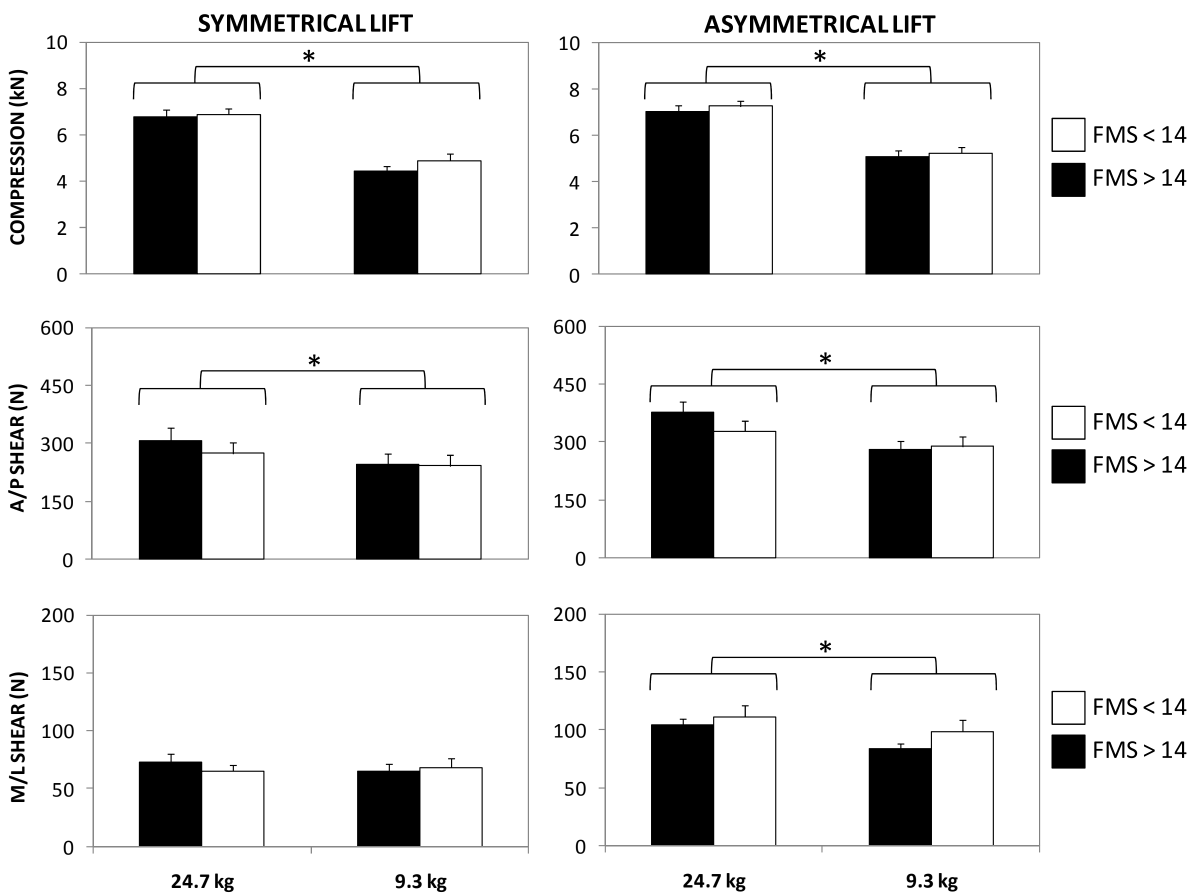

Figure 5.4. Peak L4/L5 joint compression and reaction shear forces calculated during lifting. Mean values, derived from the data of all study subjects, are plotted (N = 15 subjects/group). Error bars represent the standard error of the mean. * indicates that differences were statistically significant ($p < 0.05$).

Significantly greater peak L4/L5 joint compression ($p < 0.0001$) and A/P reaction shear ($p \leq 0.0021$) forces were documented when subjects lifted the heavier of two masses (24.7 kg vs. 9.3 kg) (Figure 5.4). However, a statistically significant effect of mass lifted on peak L4/L5 M/L reaction shear forces was only observed when performing asymmetrical lifts ($p = 0.0094$).

At the time when peak compressive forces were borne by the low-backs of subjects (i.e., at PEAK frame), the orientation of the lumbar spine (thorax with respect to pelvis) was not different between high- and low-scorers about the lateral bend ($p \geq 0.4215$), axial twist ($p \geq 0.2734$), or flexion/extension ($p \geq 0.1354$) axes.

5.4. Discussion

Influenced by the research conducted by Kiesel and colleagues (Kiesel et al. 2006; Kiesel et al. 2007; Kiesel et al. 2009), the aim of this study was to compare the peak low-back loading response to lifting between subjects who achieved a composite FMS score of greater or less than 14. It was hypothesized that subjects belonging to the high-scoring group (FMS > 14) would load their low-backs differently when lifting than would size-matched low-scorers (FMS < 14). Results obtained did not support this hypothesis, as the peak low-back loading response was not different between the high- and low-scorers. Although pre-participation movement screening has been proposed a way to predict who might sustain musculoskeletal injuries when performing physically demanding occupational or athletic tasks, results of this study do not support the notion that the potential of sustaining an acute low-back injury when lifting could be predicted based on scoring above or below 14 on the FMS.

Several limitations of this study must be considered in the interpretation of the results. First, as discussed in more detail previously (Chapter 4) and again in Chapter 7, the biomechanical modeling approach employed was unable to account for any between-group differences that may have existed in trunk muscle activation patterns. Consequently, it is possible that between-group differences in low-back loading responses went undetected, though peak low-back loading estimates derived using more sophisticated optimization- and EMG-assisted musculoskeletal models are consistent with those generated using methods comparable to the approach used in this study (Gagnon et al. 2001; Fathallah et al. 1999).

A second and arguably more important limitation to consider is that a previously reported injury prediction cut-point (composite FMS score <> 14) was used to distinguish between individuals who did and did not demonstrate a general ability or preference to move and control their bodies in ways regarded as “normal” or “desirable”. The FMS is promoted as a tool to appraise general movement behaviour and is based on the notion that individuals *should* be able to move freely, symmetrically, and without pain (Cook 2003; Cook et al. 2010). Although FMS grading criteria are indeed congruent with (and in many cases based on) current research- and clinically-informed recommendations and opinions offered in respected texts (e.g., Kendall et al. 2010; McGill 2007; Sahrmann 2002; Sahrmann 2010), the composite FMS score has not yet been evaluated with respect to its ability to measure “normal” or “desirable” movement *per se*. Rather, Kiesel and colleagues (Kiesel et al. 2007; Keisel et al. 2009) used receiver operating characteristic curves to identify the composite score that best distinguished (by maximizing classification sensitivity and specificity) between individuals who did or did not miss athletic or occupational training time due to musculoskeletal complaints. Although links between musculoskeletal injury and joint mobility restrictions, bilateral asymmetries, postural control deficits, and pain are certainly plausible, it does not necessarily follow that an indicator of injury risk can be used as a valid surrogate measure for such characteristics. FMS scores might also or instead reflect aspects of some other underlying processes or constructs which may or may not be related to these undesirable personal movement qualities; performers’ attention and perception could also conceivably influence

FMS task execution, for example. Given this limitation, the negative findings of this study do not necessarily suggest that a general whole-body movement screen could not be used for the purpose proposed here; results could be alternatively interpreted to suggest that further investigation is warranted to better administer and interpret the FMS if intended to serve such purposes.

Apart from FMS measurement validity issues introduced above, an additional issue must be considered with respect to the dichotomization of study subjects into low- and high-scoring groups. In pain-free individuals, composite FMS scores less than 21 (a perfect score) indicate that during at least one FMS task, performers exhibit movements different from those deemed normal or desirable. If the argument is granted that such deviations are indeed abnormal and perhaps unfavourable then, strictly speaking, any composite FMS score less than 21 could be considered indicative of atypical/undesirable movement behaviour. Because there were no perfect scorers in this study, it was not possible to compare the low-back loading response to lifting between individuals who did (FMS < 21) and did not (FMS = 21) exhibit atypical/undesirable movement behaviours. Instead, subjects were categorized as low- or high-scorers using a composite FMS score of 14 based on the most objectively determined and clinically meaningful FMS cut-point reported to date (O'Connor et al. 2011; Kiesel et al. 2006; Kiesel et al. 2007; Kiesel et al. 2009; Raleigh et al. 2010). Results of research conducted on a large cohort of Marine officer candidates (N = 934) also support using a composite FMS score of 14 as a cut-point, as injury-related attrition rates were two times lower in candidates who scored above this threshold (FMS > 14) than they were in those who scored 14 or less (Raleigh et al. 2010). In a study examining the relationship between composite FMS scores and injury history and lost work time in a sample of 433 full duty firefighters, Peate et al. (2007) indicated that a higher cut-point (FMS = 17) best distinguished between individuals with and without previous injuries. However, more recent analyses of a subset of the Marine officer candidate data collected by Raleigh et al. (2010) suggest that it may be inappropriate to dichotomize FMS data in this way given that the relative risk of reporting injuries during training was greater in both low-scoring (FMS ≤ 14) and very high-scoring (FMS ≥ 18) candidates (O'Connor et al. 2011). It is therefore possible that a different outcome would have resulted in this study if different FMS cut-points were implemented, but it was not possible to test cut-points reported Peate et al. (2007) or O'Connor et al. (2011) given that only 3 out of 60 subjects in this study achieved a composite FMS score of 17 or greater.

It is important to note that other studies have revealed no relationships between composite FMS scores and musculoskeletal complaints in college (American) football players (Krackow 2001), high school basketball players (Sorenson 2009), firefighter candidates (Burton 2006), or recreational runners (Hoover et al. 2008). Perhaps activity- or population-specific cut-points would yield more consistent findings, or maybe grading particular FMS tasks (e.g., deep squat) would be better suited to predict how individuals would move and load their musculoskeletal system during similar job duties (e.g., lifting) or athletic manoeuvres (e.g., jumping). However, it was not possible to directly address such questions in this study *a posteriori* given the relatively small and homogenous group of subjects tested.

Despite the fact that the study hypothesis was rejected, results of this investigation do not exclude the possibility that pre-participation movement screening could be used to identify personal qualities which have the capacity to influence injury potential at work. In athletic populations, it has been shown that individuals who exhibit postural control deficits during pre-participation testing, for example, are more likely to suffer injuries than are individuals without such deficits (Hewett et al. 2005; Zazulak et al. 2007). Based on sound clinical and biomechanical arguments to explain such findings (e.g., Hewett and Myer 2011; Powers 2010; Zazuluk et al. 2008), it is reasonable to suggest that pre-participation movement screening could also be used to predict who might get injured when performing physically demanding jobs. As indicated in the discussion points above, the challenge is to devise movement screening tasks which are capable of identifying, with high levels of specificity and sensitivity, factors known or hypothesized to influence injury potential. In this regard, there remain many opportunities to conduct future research aimed at bettering the implementation and interpretation of existing or future movement screens (i.e., validation studies); results of literature reviewed above indicate that such research is warranted.

5.5. Conclusions

Using the previously established injury prediction threshold value of 14, the composite FMS score did not indicate who might acutely overload their low-backs when lifting at work. Despite the negative findings of this investigation, future attempts to modify or reinterpret FMS scoring are justified given that several previous studies have revealed links between composite FMS scores and musculoskeletal complaints.

CHAPTER 6

Movement- vs. Fitness-Centric Exercise –

Firefighter Fitness, Whole-Body Movement Qualities, and Occupational Low-Back Loading Outcomes

CHAPTER 6

Movement- vs. Fitness-Centric Exercise – Firefighter Fitness, Whole-Body Movement Qualities, and Occupational Low-Back Loading Outcomes

Summary

Background: The impact of exercise on firefighter job performance and cardiorespiratory fitness has been studied extensively, but its effect on musculoskeletal loading remains less understood. The aim of this study was to compare various physical fitness, general movement quality, and low-back loading outcomes between groups of firefighters who completed fitness- or movement-centric exercise.

Methods: Career firefighters volunteered to participate and were randomly assigned to a control (CON), fitness-centric exercise (FIT), or movement-centric exercise (MOV) group. Before and after 12 weeks of exercise, subjects performed a physical fitness test battery, the Functional Movement Screen™ (FMS), and laboratory-simulated firefighting tasks during which peak L4/L5 joint compression and reaction shear forces were quantified using a dynamic biomechanical model.

Results: FIT and MOV subjects exhibited statistically significant improvements in nearly all measures of physical fitness (i.e., body composition, cardiorespiratory capacity, muscular strength, power, endurance, and flexibility), but it was not clear that either exercise program consistently altered general movement abilities, preferences, or peak low-back loading responses to simulated job demands.

Conclusions: Firefighters who are physically fit are better able to perform essential job duties and avoid cardiorespiratory failure, but improving physical fitness over a 12-week period may not necessarily reduce occupational low-back loading demands. More research is needed to better understand how individuals adapt to exercise and what impact these adaptations have on movement behaviour, low-back loading, and hypothesized injury potential.

6.1. Introduction

Given the strenuous and inherently hazardous nature of their work, firefighters are encouraged to enhance and maintain their physical fitness through exercise. Adaptations to exercise are believed to improve on-the-job performance and reduce the risk of cardiorespiratory failure and musculoskeletal injuries (Smith 2011). There have been previous studies examining the influence of exercise on measures of firefighter job-specific performance (Peterson et al. 2008) and cardiorespiratory fitness (Roberts et al. 2002; Throne et al. 2000), and it has been shown that firefighters who are more physically fit report fewer and less costly low-back injuries than do firefighters who are less physically fit (Cady et al. 1979; Cady et al. 1985). However, the impact of exercise on biomechanical variables – particularly those associated with work-related low-back injury and pain reporting (e.g., low-back loading patterns) – has been unexplored in this population. Without such information, it is difficult to devise and objectively evaluate the effectiveness of exercise-based low-back injury prevention strategies for firefighters because the link between physical fitness improvements, occupational low-back loading demands, and injury potential remains unknown.

There are several ways in which adaptations to exercise could conceivably reduce the likelihood of sustaining musculoskeletal injuries. Because the structure, composition, and quantity of bone, ligament, tendon, and skeletal muscle tissue vary in response to mechanical stimuli (Whiting and Zernicke 2008), appropriately designed exercise can conceivably cause musculoskeletal tissues to grow and remodel, making them less likely to sustain damage when loaded . Exercise could also impact habitual patterns of coordination and control by altering other inherent structural or functional (physiological and psychological) personal movement constraints such as: body segment mass-inertial characteristics (via body composition changes); flexibility and joint mobility; mechanical, electrical, and metabolic functioning of movement system components and their interactions ; cardiorespiratory fitness; perception-action response patterns; intentions ; etc.. Such changes can influence how individuals consciously or subconsciously interact with their environment when engaged in physical activity (i.e., their movement behaviour) and resulting movement strategies can modify the relationship between imposed demands (applied musculoskeletal load) and the capacity to withstand imposed demands (musculoskeletal load tolerance). Viewed from this perspective, exercise can thus reduce musculoskeletal injury potential by eliciting adaptations that cause individuals to move and activate their muscles in ways that increase their “margin of safety” (McGill 2009).

To date, there have been no known attempts to study the impact of exercise on occupational low-back loading demands of firefighters. In athletic populations, it has been demonstrated that exercise intended to alter patterns of movement coordination and control can influence sport-related musculoskeletal loading and injury risk measures (Greska et al. 2011; Myer et al. 2007). However, the same measures are not always influenced by exercise designed primarily to improve general physical fitness characteristics (Herman et al. 2008; McGinn et al. 2006; Trowbridge et al. 2005; Willy and Davis 2011). Given that both types of exercise, movement- and fitness-centric, have been shown to yield positive athletic and occupational task performance outcomes (Peterson et al. 2008; Myer et al. 2005; Myer et al. 2006), it could be argued that firefighters would be better served by adopting a movement-centric approach to exercise if the goal is to enhance and maintain work performance capabilities and low-back durability. However, this notion has yet to be tested.

The objective of this study was to compare physical fitness, general movement quality, and occupational low-back loading outcomes between groups of firefighters who completed 12 weeks of movement- or fitness-centric exercise. It was hypothesized that between-group differences in exercise outcomes would result and that differential outcomes could be used to justify one exercise-based low-back injury prevention approach over the other.

6.2. Methods

Experimental Overview

A battery of fitness, general movement quality, and low-back biomechanical measures were made before and after two groups of male firefighters completed different 12-week exercise programs. Exercise programs differed based on whether the primary emphasis (via coaching cues and exercise selection) was placed on improving physical fitness (body composition, cardiorespiratory capacity, muscular strength, power, endurance, and flexibility) or on enhancing general movement aptitude through exercise and education during training sessions (e.g., emphasis on controlling lumbar spine posture while lifting, pushing, pulling, etc.). Pre- and post-training measures of physical fitness and general movement quality were compared, as were peak L4/L5 joint compression and reaction shear forces calculated during laboratory-simulated firefighting tasks performed before and after completing 12 weeks of exercise.

Subjects and Group Assignments

Sixty male members of the Pensacola Fire Department (Pensacola, FL, United States) volunteered to participate. Subjects were free of any activity-limiting health conditions (e.g., chronic musculoskeletal pain, cardiorespiratory disorders, etc.) that would have impacted their ability to participate without risk of exacerbating signs and symptoms. A questionnaire was used to acquire information about the general health of subjects, and subjects also signed informed consent documents that were approved by the University of Waterloo's Office of Research Ethics, the Baptist Hospital Institutional Review Board, and the City of Pensacola.

Subjects were assigned to one of three groups: 1) fitness-centric exercise group (FIT); 2) movement-centric exercise group (MOV); or 3) control group (CON). It was attempted to randomly assign subjects to groups, however a number of concessions had to be made to accommodate scheduling conflicts (many firefighters had second jobs) and to satisfy requests from individuals who needed or wanted to car-pool (round-trip commute times for some subjects exceeded 60 minutes). In cases where a subject needed to be re-assigned (to accommodate scheduling or commuting conflicts), a switch was made with a subject from a different group who did not indicate any conflicts. No restrictions were placed on members of the control group, as they were simply instructed to continue living as they normally would.

Experimental Protocol and Data Collection

Before and after completing the 12-week exercise programs, subjects scheduled three separate pre-training data collection sessions – conducted on different days – wherein tests of physical fitness, general movement quality, or simulated job tasks were conducted. After pre-training collection sessions were completed (typically within a two-week period) subjects scheduled their first “workout” (i.e., exercise training session). The first workout was to be completed within 3 to 7 days following their final pre-training data collection session to permit adequate recovery time and to allow coaches to gradually integrate of new trainees into the exercise program without impeding the progress of subjects who had already started. An additional benefit of this “waterfall-like” approach was that it allowed for post-training fitness, movement quality, and biomechanical data collection sessions to be initiated within a 3- to 7-day time period immediately following conclusion of the 12-week exercise program. In this way, it was hypothesized that adequate recovery time would be provided to subjects (between their final workout and their first post-training data collection session) while minimizing the attenuation of any exercise adaptations incurred. Pre- and post-training fitness, movement, and biomechanical data collection sessions were performed in exactly the same way and within a similar timeframe so that the impact of training could be examined.

In one data collection session, subjects completed a fitness test battery. The order in which fitness tests were performed was randomized between subjects, but within-subject test order remained fixed in the pre- and post-training data collection sessions. Tests were selected to measure multiple characteristics of physical fitness (i.e., body composition, cardiorespiratory capacity, muscular strength, power, endurance, and flexibility), and included the following:

Subcutaneous Body Fat. An experienced research assistant used standard calipers to make skinfold measurements (mm) from the following seven sites:

i. Chest – diagonal fold, one-third of the way between upper armpit and nipple;
ii. Abdominal – vertical fold, 2.54 cm to the right of navel
iii. Thigh – vertical fold, midway between knee cap and top of thigh
iv. Tricep – vertical fold, midway between elbow and shoulder
v. Subscapular – diagonal fold, directly below shoulder blade
vi. Suprailiac – diagonal fold, directly above iliac crest
vii. Midaxillary – horizontal fold, directly below armpit

Total body fat percentage for each subjects was estimated based on the following inputs (Jackson and Pollock 1978): age; mass; height; gender; sum of skinfold measurements.

Gerkin Treadmill Protocol. Subjects performed the Gerkin treadmill workload protocol (Gerkin et al. 1997) to appraise their cardiorespiratory capacity, a standardized submaximal test. The protocol began with a 3-minute warm-up during which subjects walked on a motorized treadmill at a constant speed of 3 miles/hour and 0% grade. After the warm-up, treadmill speed was increased to 4.5 miles/hour for 60 seconds. Treadmill speed and grade were then alternately increased every minute in 0.5 miles/hour and 2% increments, respectively, until subjects were unable or unwilling to continue. Total time to test completion was recorded (in seconds), and maximum oxygen consumption (VO_2 max) was estimated based on equations derived by Tierney et al. (2010).

Push-Ups. Subjects performed push-ups until they were unable or unwilling to continue whilst demonstrating proper form. Initial posture was standardized by instructing subjects to place their hands under their shoulders and they were instructed to control lumbar spine posture (i.e., they were told to maintain a "tight" midsection). Any repetitions during which elbows did not fully extend or the chest did not touch pads located beneath them were not included in the total (Figure 6.1). The maximum number of push-ups that could be performed continuously and with proper form was recorded.

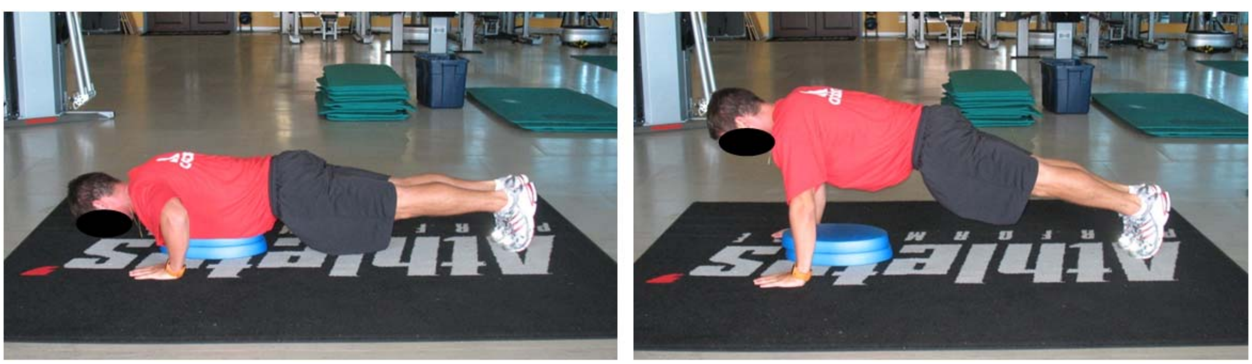

Figure 6.1. Push-up test.

Trunk Muscle Endurance. Isometric torso flexion, extension, and lateral bend exertions were performed to test trunk muscle endurance. As described in McGill et al. (2010), exertions were performed in the "plank" (flexion endurance), "Biering-Sørensen" (extension endurance), and "side bridge" (lateral bend) positions (Figure 6.2). Subjects were instructed to maintain the isometric exertions for as long as possible, and the total time to task failure was recorded (in seconds). Consistent with criteria used by McGill et al. (2010), the task was terminated when subjects were unable or unwilling to preserve their posture after one verbal warning.

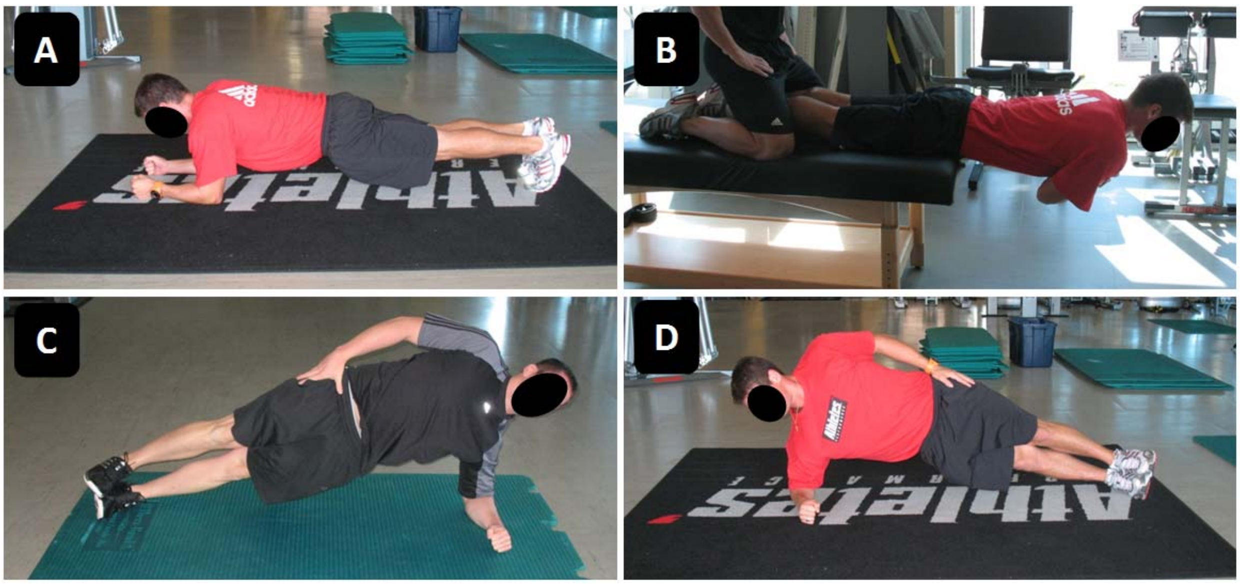

Figure 6.2. Trunk muscle endurance tests. A) Plank – test of trunk flexion endurance; B) Biering-Sørensen – test of trunk extension endurance; C) Left Side Bridge – test of left lateral bend endurance; D) Right Side Bridge – test of right lateral bend endurance.

Upper Body Power. While seated in a Keiser® AIR250 chest press machine (Keiser Corporation, Fresno, CA, United States), subjects executed maximum-speed upper body bilateral pressing exertions at five pneumatically-controlled load settings: 30, 50, 70, 90, 110 lb (Figure 6.3). Five trials at each load setting were performed. Approximately 10 s rest was provided between exertions, and 60 s rest was provided between loads. Peak power display settings were recorded (in Watts) for each trial and the median value (over 5 trials) was used in the statistical analyses. If the torso did not maintain contact with the backrest throughout the pressing exertion, the recordings were discarded and the test was repeated.

Figure 6.3. Upper body power output test using the Keiser® AIR250 chest press machine.

Lower Body Power. Two tests of lower body power were performed. Using a Keiser® AIR300 squat machine (Keiser Corporation, Fresno, CA, United States), subjects executed maximum-speed bilateral squatting exertions at five pneumatically-controlled load settings: 40, 60, 90, 120, 150 lb (Figure 6.4). Five trials at each load setting were performed. Approximately 10 s rest was provided between exertions, and 60 s rest was provided between loads. Peak power display settings were recorded (in Watts) for each trial and the median value (over 5 trials) was used in the statistical analyses. The initial squat posture was controlled by adjusting the Keiser® machine such that knees were flexed to 90 degrees (measured using a goniometer). Subjects were asked to produce maximum efforts, but were not permitted to jump (i.e., their feet could not leave the platform).

Figure 6.4. Lower body power output test using the Keiser® AIR300 squat machine.

As a second measure of lower body power, subjects performed maximum-effort vertical jump tests. Maximum standing bilateral reach height was first recorded, and the maximum jump height was obtained using a Vertec Vertical-Jump Tester (Gill Athletics, Champaign, IL, United States) (Figure 6.5). The maximum height achieved among three counter-movement jump trials was recorded. Measurements were made in inches and subsequently converted to centimeters. Full recovery was permitted between jumps, and vanes of the Vertec device were not reset between exertions to motivate the subjects to better their previous performance.

Figure 6.5. Vertical jump test for lower body power output using the Vertec Vertical-Jump Tester.

Grip Strength. Right- and left-hand grip strength was measured by instructing subjects to maximally squeeze a hand dynamometer. Tests were performed in a seated position (90 degrees of knee flexion), with arms vertically oriented and the test-side elbow flexed to 90 degrees (Figure 6.6). No movement (with respect to the starting position) was permitted, and subjects were encouraged to execute maximum ramped contractions. The peak value achieved among three trials was recorded (in kilograms) for inclusion in statistical analyses. Full recovery was provided between trials.

Figure 6.6. Right-hand grip strength test.

Sit-and-Reach. Without wearing shoes, subjects performed three maximal sit-and-reach trials using a standard test box (Figure 6.7). Only measurements resulting from slow, controlled symmetrical movements were recorded (i.e., no bouncing or twisting was permitted). The maximum value achieved (in centimeters) among three trials was used in statistical analyses.

Figure 6.7. Sit-and-reach test.

In a second data collection session, general movement quality was measured using the Functional Movement Screen™ (FMS). As outlined in the previous chapter (and detailed in Appendix II), the FMS consists of seven tests which are ranked based on the performers ability to execute general whole-body movement tasks without experiencing pain and without exhibiting compensatory movement behaviours that are hypothesized to represent deficiencies or limitations in movement coordination and control. As such, the FMS was included in this study to determine if exercise training altered the general ability to move freely, symmetrically, and without pain. Exactly as described in the previous chapter, subjects were videotaped while performing FMS tasks and scoring was conducted off-line by a single trained observer on the basis of the video recordings.

In a third data collection session conducted in a biomechanics laboratory, subjects performed a test battery consisting of general whole-body pushing, pulling, and lifting tasks together with several simulated firefighter-specific tasks. Drawing on the knowledge and experience of PFD training officers and taking practical considerations into account, tasks were selected and designed to reflect a range of physically demanding tasks that firefighters perform at the station (e.g., manual handling activities) or during fireground operations (e.g., structural fire suppression). Simulated manual handing duties consisted of symmetrical lift, asymmetrical lift, unilateral push, and unilateral pull tasks, whereas simulated fireground duties consisted of ceiling breach, ceiling pull, forcible entry, overhead chop, and hose pull tasks. When performing the fireground tasks, a weighted vest (22.7 kg) was worn by subjects to simulate the mass of a self-contained breathing apparatus and turnout gear. The weighted vest was not worn when performing the material handling tasks. Laboratory task descriptions are as follows:

Symmetrical Lift. From a relaxed upright standing posture, subjects were asked to bend forward, grasp, and lift a 24.7 kg crate from the floor directly in front of them (Figure 6.8). Subjects were instructed to "lift as naturally as possible" provided that their feet remained in-contact with on the force platforms. Three trials were collected, during which single lifting exertions were performed.

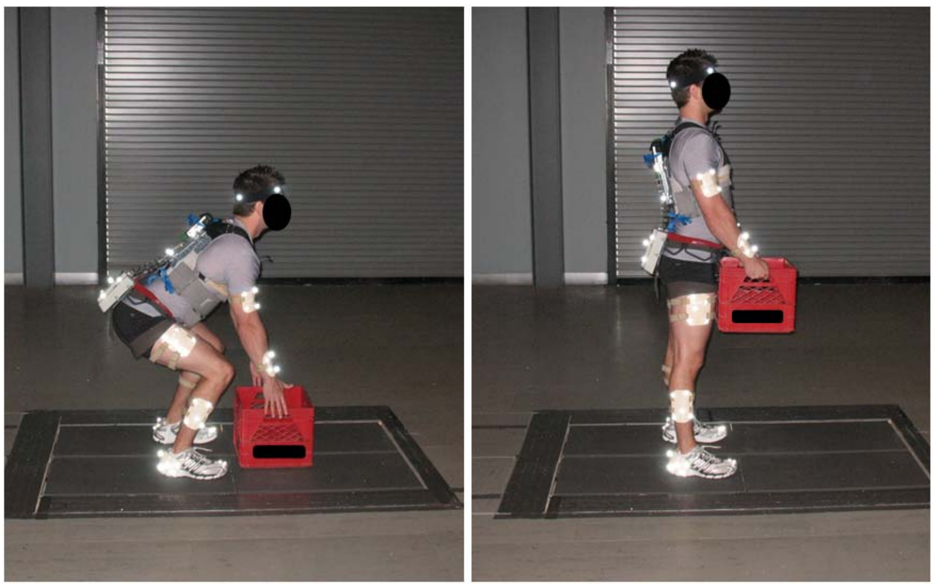

Figure 6.8. Symmetric lifting task (SYMM).

Asymmetrical Lift. Subjects were asked to lift from the floor a 24.7 kg crate that was positioned at approximately 45 degrees with respect to the mid-sagittal plane (Figure 6.9). Initial foot position was not strictly controlled, as subjects were free to execute the lift as they "naturally would". However, subjects were instructed to "keep their feet on the force platforms". Three trials were collected, during which single lifting exertions were performed.

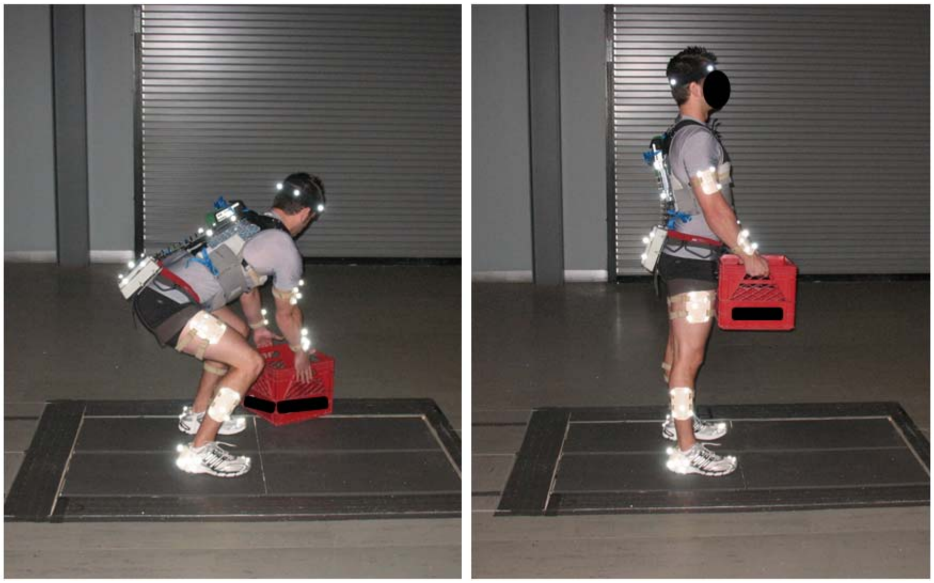

Figure 6.9. Asymmetrical lifting task (ASYM).

Unilateral Push. With their feet arranged in a split-stance configuration on the force platforms (i.e., left foot forward, right foot back), subjects were asked to perform a resisted pushing motion with their right arm (Figure 6.10). A handle, attached in-series to a pneumatic cable resistance machine (Figure 6.11), was held in the right hand at the right side of the body, pushed directly forward until the right elbow was fully extended, and then returned to the starting position. Given the instruction to perform "as naturally as possible", speed was self-selected. Measured cable resistance was 96 N. Three trials were collected, during which single pushing exertions were performed.

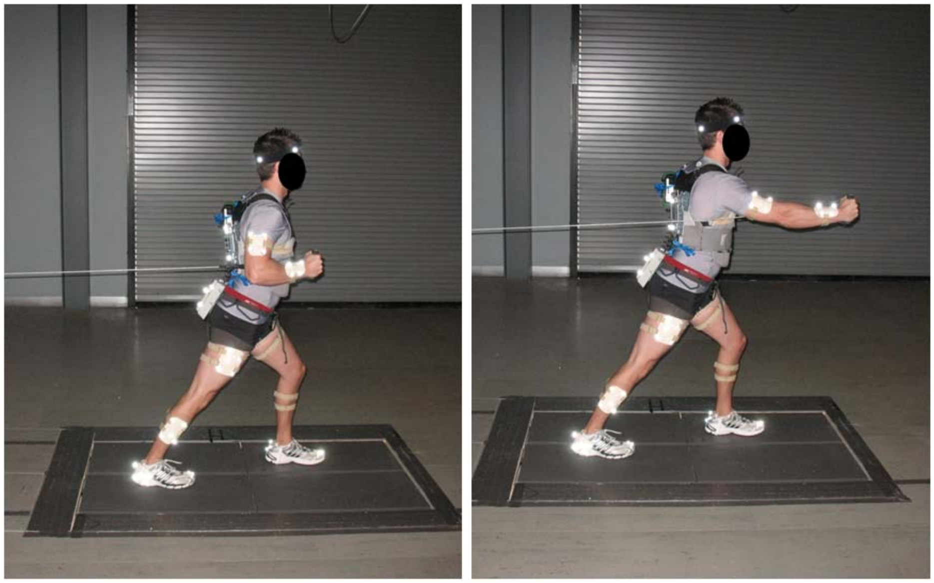

Figure 6.10. Unilateral push task (PUSH).

Figure 6.11. Keiser® “Functional Trainer” pneumatic cable resistance machine.

Unilateral Pull. A resisted right-handed pulling motion was also performed “as naturally as possible” with the feet arranged in a split-stance configuration on the force platforms (i.e., left foot forward, right foot back). With their right elbow fully extended, subjects grasped the handle with their right hand, pulled the cable directly to their right side, and then returned to the starting position (Figure 6.12). Measured cable resistance was 133 N. Three trials were collected, during which single pulling exertions were performed.

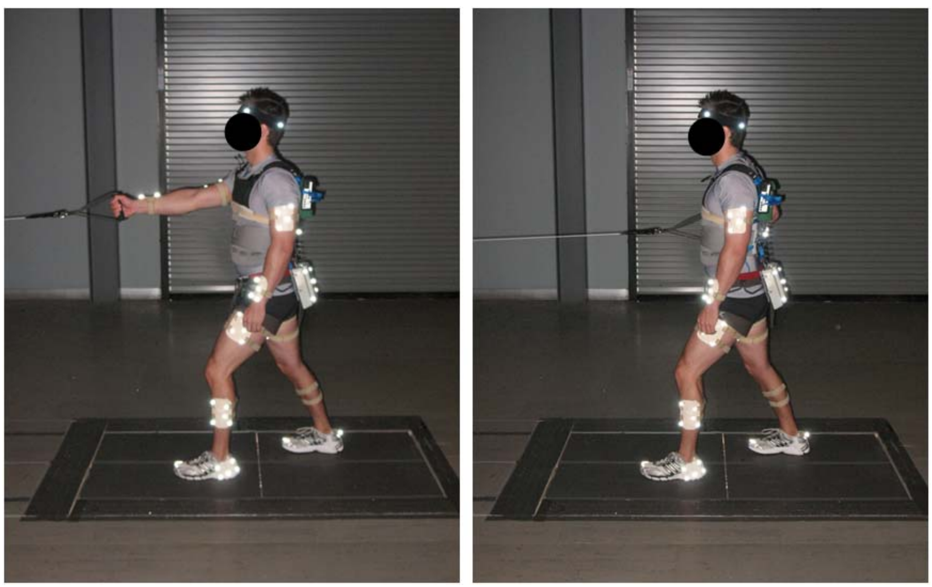

Figure 6.12. Unilateral pull task (PULL).

Ceiling Breach. By pushing a pike pole overhead against resistance, subjects simulated the act of breaking-though a ceiling to inspect for fire extension (Figure 6.13). Resistance was applied through a pneumatically controlled system of cable and pulleys. Measured cable resistance was 135 N. Within a single trial, five repetitions were performed at a self-selected pace. Two trials were collected.

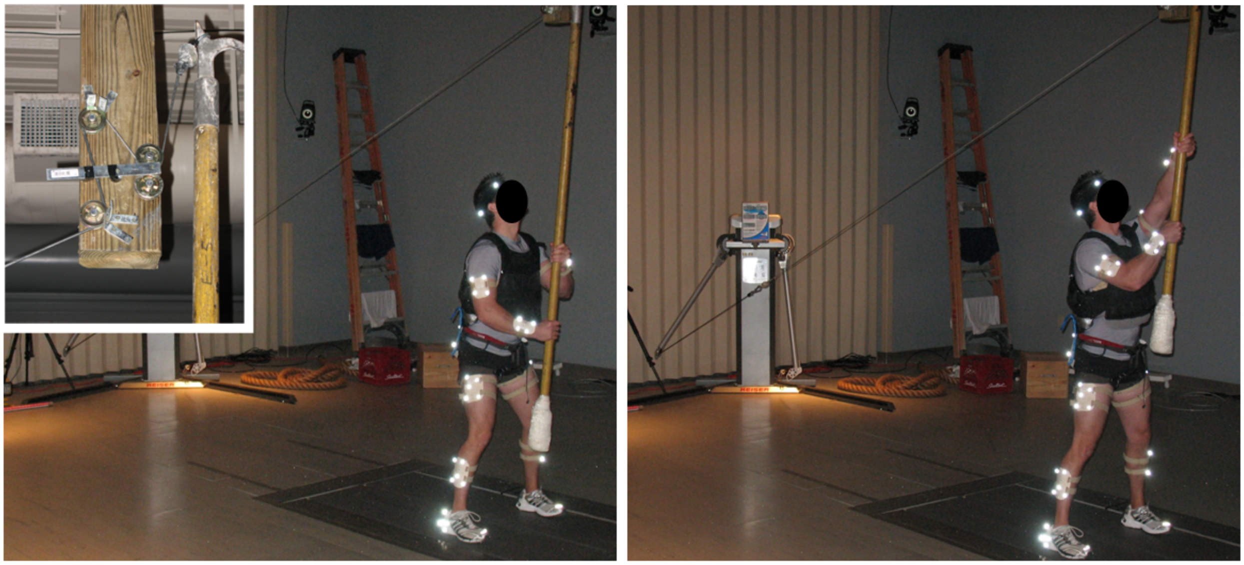

Figure 6.13. Ceiling breach task (CBRC).

Ceiling Pull. To simulate the act of removing ceiling to check for fire extension, a pike pole (to which a resistance cable was again connected through a pulley system) was pulled downward (Figure 6.14). Measured cable resistance was 219 N. Within a single trial, five repetitions were performed at a self-selected pace. Two trials were collected.

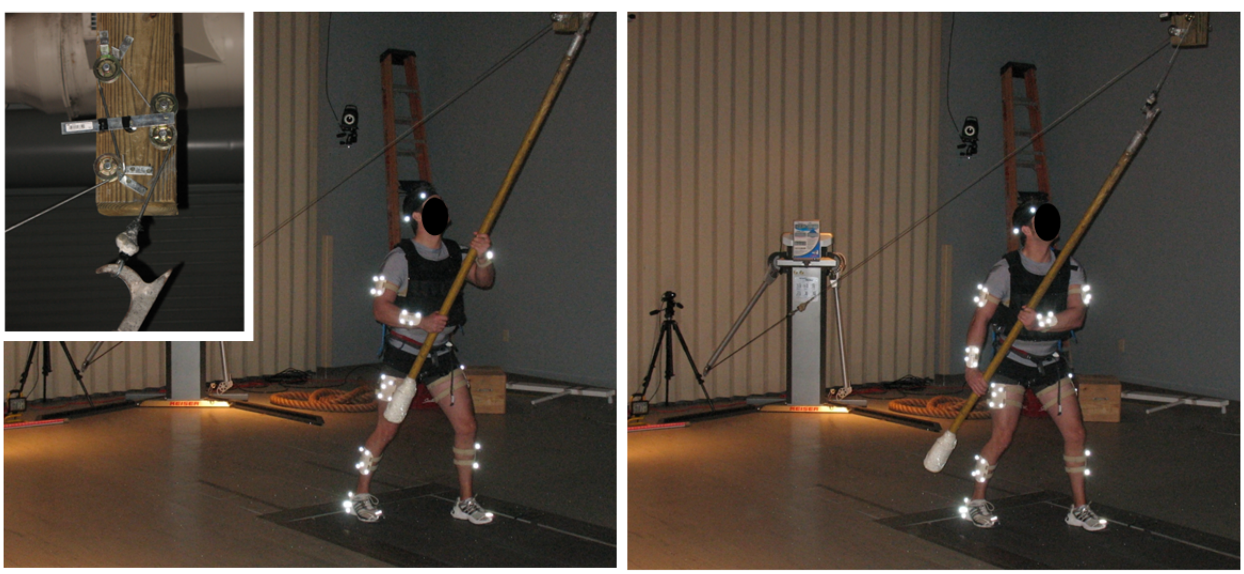

Figure 6.14. Ceiling pull task (CPUL).

Forcible Entry. Intended to simulate the act of entering a building through a lodged door or wall, subjects consecutively struck a 45.4 kg heavy bag with a 4.5 kg sledgehammer (Figure 6.15). Subjects were asked to ensure that their feet were positioned on the force platforms before starting, but after initiation, they were instructed to focus solely on task execution. (Foot positioning was monitored by a research assistant; trials were recollected if subjects stepped-off the force platforms.) Within a single trial, five strikes were performed at a self-selected pace. Two trials during which feet remained on the force platforms were saved for analyses.

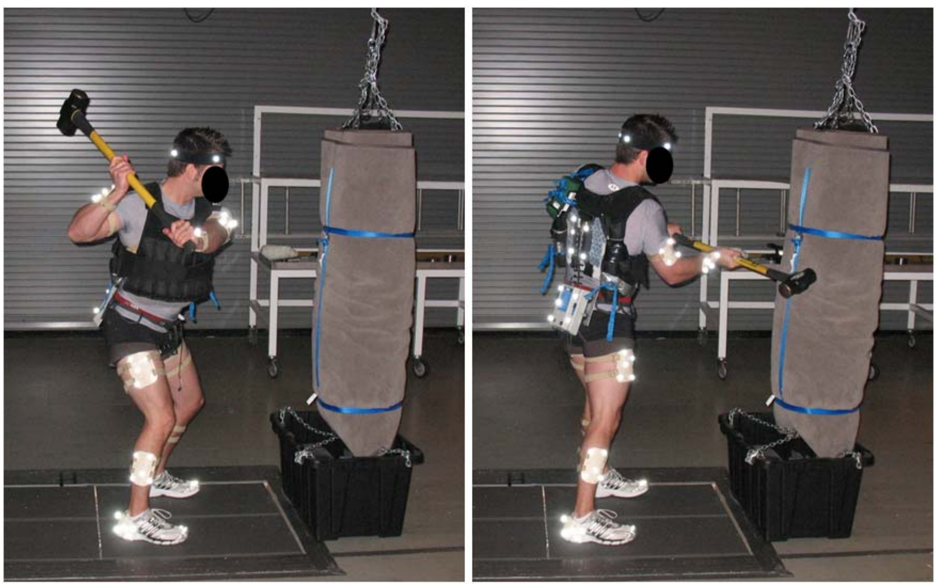

Figure 6.15. Forcible entry task (FENT).

Overhead Chop. Also using the 4.5 kg sledgehammer, subjects simulated the act of infiltrating a structure by chopping downward from an overhead position (Figure 6.16). Subjects were again directed to focus on task execution, as a research assistant would monitor their foot positioning. Within a single trial, five consecutive chops were performed at a self-selected pace. Two trials during which feet remained on the force platforms were saved for analyses.

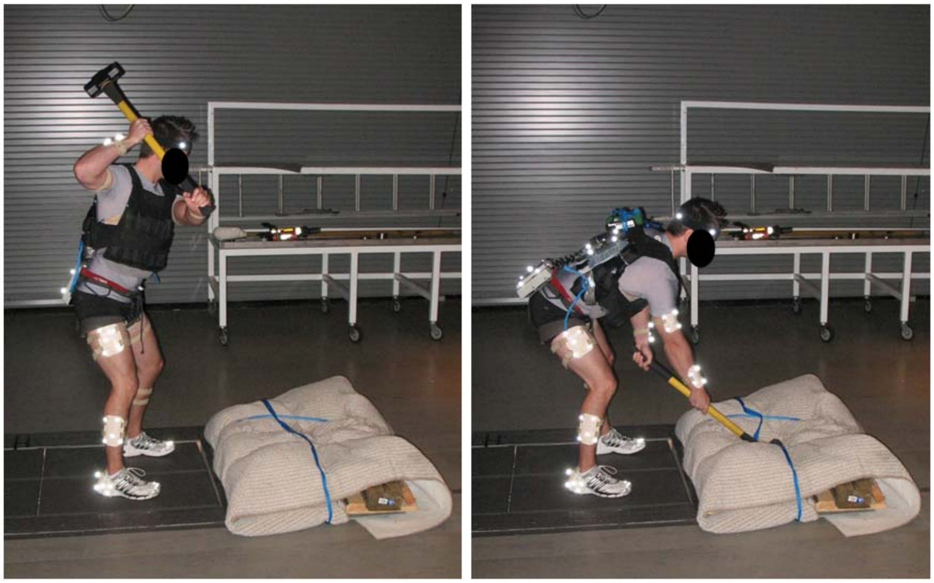

Figure 6.16. Overhead chop task (CHOP).

Hose Pull. A rope, connected to the pneumatic resistance cable machine, was pulled by the subjects in a hand-over-hand fashion to simulate pulling a charged hose (Figure 6.17). Measured cable resistance was 133 N. Within a single trial, a minimum of three pulls (e.g., right-left-right) were performed at a self-selected pace. Three trials were collected.

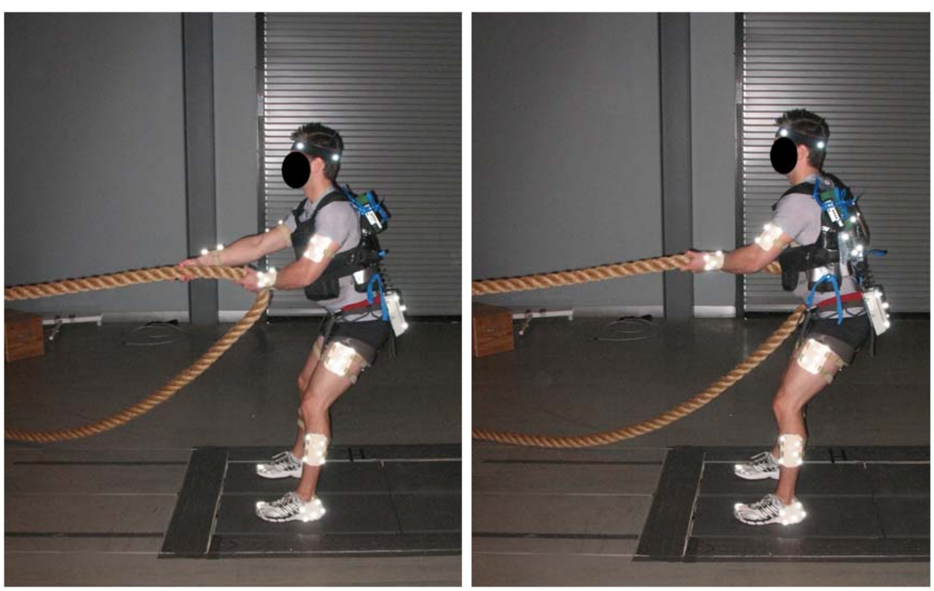

Figure 6.17. Hose pull task (HPUL).

In the biomechanics laboratory, three-dimensional kinematics of the trunk, pelvis, and lower extremities were captured at 160 Hz using a 10-camera Vicon motion capture system (Vicon, Oxford, United Kingdom), and four ground-mounted force platforms were used to measure ground reaction forces and moments at a sample rate of 2400 Hz (Bertec Corporation, Columbus, OH, United States) (Figure 6.18). Vicon Nexus Motion Capture software (Version 1.5, Vicon, Oxford, United Kingdom) was used to synchronize, capture, and store biomechanical data for post-processing. As described in previous chapters, “rigid body” marker clusters were secured to the trunk, pelvis, thighs, shanks, and feet of subjects to track the motion of body segments during experimental trials. Additional markers were included in a static calibration trial to generate anatomically-meaningful segment-fixed coordinate systems (as depicted in the previous chapter: Figure 5.2). Using Visual3D™ software (Version 4, C-Motion, Inc., Germantown, MD, United States) in the same way described previously (Chapters 3, 4 and 5), body segment kinematics and ground reaction forces were used to compute net three-dimensional L4/L5 joint reaction forces and moments via inverse dynamics. Orthogonal components of the net L4/L5 joint moment were input into a polynomial equation to yield estimates of the L4/L5 “bone-on-bone” compression force (McGill et al. 1996b). Peak L4/L5 joint compression and reaction shear forces were extracted from each laboratory task trial, and the absolute value of the lumbar spine angle was recorded at the instant when L4/L5 compression force was greatest. The lumbar spine orientation calculated in a relaxed upright standing trial was considered to represent zero degrees about the flexion/extension, lateral bend, and axial twist axes.

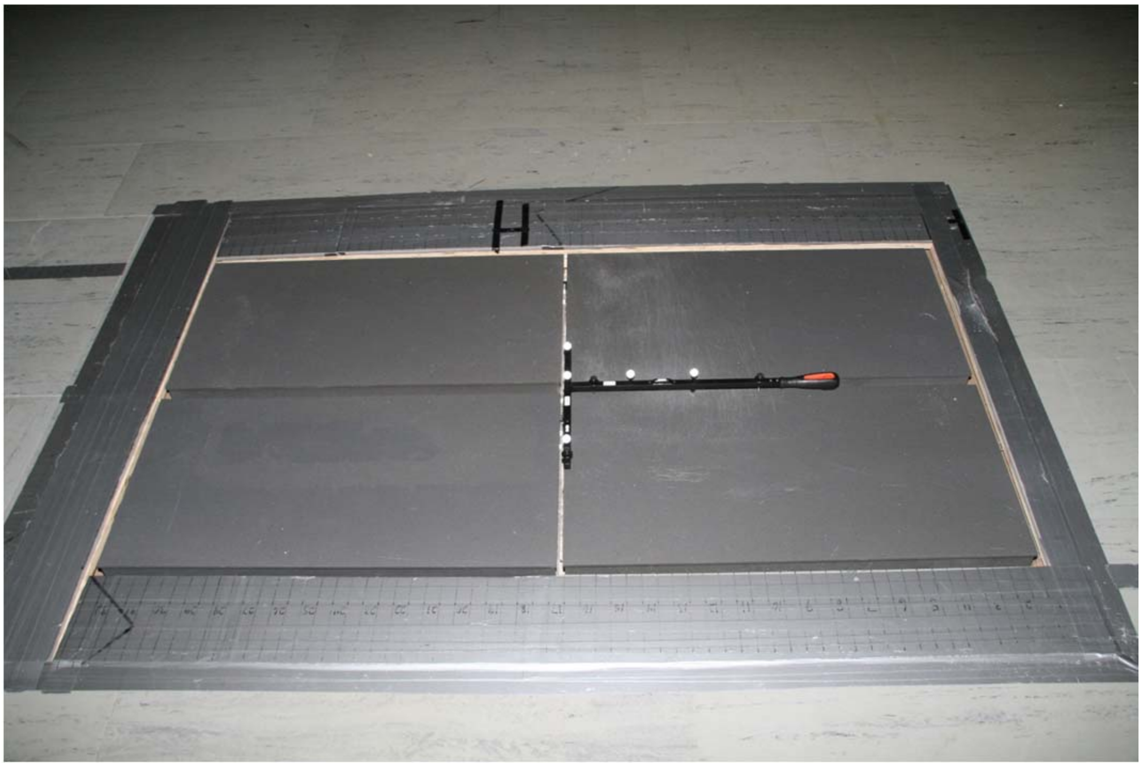

Figure 6.18. Layout of the force platforms.

Exercise Programs

Exercise programs differed based on whether the primary emphasis was placed on improving measures of physical fitness (e.g., muscular strength, endurance, flexibility, etc.) or whether the primary emphasis was placed on improving patterns of movement coordination and control. To achieve this, the selection, order, and progression of exercises differed between the programs as did the focus and intent of coaching instruction and feedback. Accredited strength and conditioning coaches administered the exercise programs via small group instruction (i.e., 3 to 6 firefighters/coach). Coaches were experienced (i.e., they had been coaching for over 2 years) and their college-level education in Exercise Science exposed them to the different instruction and feedback approaches used in this study. Though they understood the exercise program objectives and were provided details about the physical fitness and movement quality tests, coaches were not made aware of the laboratory-based biomechanical testing protocol to avoid biasing their instruction and feedback approaches.

The FIT program was based on a non-linear periodization scheme, crafted specifically (via exercise selection, organization, and progression) to maximize improvements in the physical fitness measures collected. Non-linear periodization programs are characterized by daily, weekly, and/or monthly fluctuations in training volume, intensity, and frequency and can be used to achieve concurrent and balanced improvements in all aspects of physical fitness (c.f., Kok et al. 2009; Peterson et al. 2008; Prestes et al. 2009). Instruction and feedback provided by the FIT program coach was intended to direct the attention of performers to the *effect* of their movements on the environment (i.e., weight lifted, number of repetitions performed, speed of execution, etc.) rather than on the movements themselves. Researchers and practitioners often refer to this type of extrinsic feedback as “external” (Wulf et al. 1998); “knowledge of results” is provided to maximize performance outcomes (Wulf et al. 2010). A general FIT program template was followed (Appendix III), but exercise intensity was individualized based on the ability of subjects to complete the prescribed number of sets, repetitions, or times. Although the FIT program coach relied heavily on motivation and encouragement to elicit maximal improvements in physical fitness, he was also free to make minor program modifications (e.g., exercise additions/substitutions) if he a felt subject was not progressing or the prescribed exercises were too advanced.

In the MOV program, exercise selection, order, and progression was also based on a non-linear periodization model because firefighter safety and work ability could have been compromised if any physical de-conditioning was permitted. In contrast to the FIT program objectives (i.e., to maximally improve physical fitness test scores), the MOV program aimed to improve physical fitness while also striving to eradicate and prevent any undesirable movement behaviours through coaching instruction (including demonstration) and verbal feedback. For example, when executing squatting, lunging, and jumping movements, some individuals are unable or choose not to prevent hip abduction and internal rotation (Cortes et al. 2007; Ford et al. 2003; Hughes et al. 2008). This behaviour might not only limit acute performance outcomes (e.g., weight lifted) and subsequent improvements in physical fitness (e.g., strength), but the potential for sustaining musculoskeletal injuries during physical activity could be increased due to a progressive weakening of vulnerable tissues (via reduced load tolerance). Such behaviour may also increase the likelihood of applying damaging tissue loads when executing squat-, lunge-, and jump-like movements in life. Similarly, when performing upper-body pushing and pulling tasks, individuals may be unable or elect not to control the position and orientation of the lumbar spine and scapulae and their performance outcomes and musculoskeletal durability could be diminished as a result. In this study, the MOV program coach used instruction and feedback to direct attention of the performers to the positions and orientations of their body segments when executing exercises based on the hypothesis that joint-sparing movement behaviours would be learned and would emerge when executing post-training simulated firefighting tasks (i.e., "transfer of training"). Providing "knowledge of performance" to trainees can be referred to as "internal" feedback (Wulf et al. 1998; Wulf et al. 2010) and is often used in clinical contexts to guide in the musculoskeletal rehabilitation process.

Instruction and feedback guidelines followed by the MOV program coach were based largely on the research results and clinical observations of Hewett, Myer, and colleagues (Hewett et al. 2007; Myer et al. 2008), Kendall et al. (2010), McGill (2006; 2007), Richardson et al. (2004), and Sahrmann (2002; 2010), but the writings of Cook et al. (2003; 2010), Boyle (2004; 2009), and Verstegen (2003; 2006) were also influential given that the aforementioned research and clinical observations are incorporated into practical guidelines for exercise prescription and progression. Since it was not feasible to strictly individualize exercise, the MOV program was generically designed to address the most common movement-related deficiencies and limitations exhibited by athletes and patients. Particular emphasis was placed on static and dynamic postural control of the lumbar spine, hips, and shoulder complex during exercise execution (e.g., Appendix IV), and joint mobility exercises were included to address commonly observed limitations (e.g., rotation through the hips, thoracic spine, and ankles). Though a general template was followed by all MOV program subjects (Appendix III), individualized instruction and feedback was provided by the MOV program coach together with any supplementary exercises deemed necessary to elicit desired movement-based adaptations. For instance, if the MOV program coach felt that a subject was not progressing or if prescribed exercises were too advanced, the coach was free to make program additions or substitutions. The MOV program coach was instructed to visually inspect for any “weak links”/“energy leaks” (i.e., uncontrolled movements) in the kinematic chain and was to base instruction and feedback on observed deficiencies. When individuals did not exhibit movement-related deficiencies, exercise was progressed until uncontrolled movements emerged. Essentially, observed patterns of movement coordination and control functioned as a “compass” to guide exercise prescription and progression within the general MOV program template.

Statistical Analyses

The impact of exercise on measures of physical fitness and low-back loading was tested using general linear models with one between-subject factor (“group”) and one within-subject factor (“time”). Mean values of the fitness and low-back loading measures calculated across (within-task) trials formed the dependent variables in the statistical analyses. Given study objectives, the primary undertaking was to identify any statistically significant group×time interaction effects (i.e., $p < 0.05$), though the “direction” and “location” of any changes was also of interest. Accordingly, when group×time interaction effects were statistically significant, a least-square means procedure with adjustments for multiple comparisons (via the Tukey method) was used. The influence of exercise on FMS scores was examined using Wilcoxon signed-rank tests. All statistical analyses were performed using SAS system software (Windows Version 9.1.3 with Service Pack 4, SAS Institute Inc., Cary, NC, United States).

6.3. Results

There were three subjects who voluntarily withdrew from the study and one subject who was unable to complete 83% of the prescribed exercise program (30/36 training sessions in 12 weeks was set *a priori* as a minimum level of compliance for study inclusion). Equipment malfunction resulted in loss of biomechanical data for an additional two subjects. Included in the results below are data from the subjects who finished the study and for which full data sets were obtained (CON = 16 subjects; FIT = 18 subjects; MOV = 20 subjects).

Physical Fitness Measures

As summarized in Table 6.1 (and in Appendix V), FIT and MOV subjects exhibited statistically significant improvements in most measures of physical fitness, but the magnitude of fitness improvements sometimes differed between the exercise groups. With the exception of select changes in upper body strength, power, and endurance, the physical fitness of CON subjects remained relatively stable over 12 weeks.

The body mass of all subjects remained constant over the course of the experiment (p = 0.5231). However, body fat percentages significantly decreased in both FIT (p = 0.0108) and MOV (p = 0.0072) subjects by an average of 1.4% body fat (Table 6.2). No differences in body composition were detected in CON subjects (p = 0.6786).

Table 6.1. Global summary of physical fitness results for subjects in the control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV). No changes in physical fitness measures over the course of the study are indicated (—) together with percentage increases (↑) and decreases (↓).

Physical Fitness Measure	**CON**	**FIT**	**MOV**
Body Mass	—	—	—
Body Fat Percentage	—	↓7.4%	↓8.1%
Treadmill Time	—	↑13%	↑9.8%
Predicted $V0_2$ max	—	↑12%	↑8.3%
Trunk Flexion Endurance	—	↑75%	↑55%
Trunk Extension Endurance	—	↑65%	↑42%
Trunk Right Lateral Bend Endurance	—	—	—
Trunk Left Lateral Bend Endurance	—	↑42%	—
Push-Ups	↑11%	↑62%	↑38%
Keiser Chest Press @ 30 lb	—	↑13%	—
Keiser Chest Press @ 50 lb	—	—	—
Keiser Chest Press @ 70 lb	↑0.1%	↑2.7%	↑8.8%
Keiser Chest Press @ 90 lb	↑1.4%	↑7.8%	↑11%
Keiser Chest Press @ 110 lb	—	↑11%	↑13%
Vertical Jump	—	↑5.2%	↑4.9%
Keiser Squat @ 40 lb	↑17%	↑54%	↑26%
Keiser Squat @ 60 lb	—	↑26%	↑18%
Keiser Squat @ 90 lb	—	↑20%	↑13%
Keiser Squat @ 120 lb	—	↑20%	↑15%
Keiser Squat @ 150 lb	—	↑16%	↑12%
Right Grip Strength	↑3.6%	↑3.1%	↑4.8%
Left Grip Strength	↑4.0%	↑4.1%	↑4.4%
Sit-and-Reach	—	—	↑22%

Table 6.2. Mean (SEM) age, height, body mass, and estimated body fat of control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects before (Pre) and following (Post) the 12-week intervention.

Group	Age, yrs	Height, cm	Body Mass, kg		Body Fat, %	
			Pre	Post	Pre	Post
CON	39 (2.2)	1.78 (0.01)	93.0 (3.9)	92.7 (3.9)	18.7 (1.8)	18.9 (1.8)
FIT	35 (2.3)	1.80 (0.01)	95.1 (3.0)	94.7 (2.9)	18.5 (1.8)	17.1 (1.5)†
MOV	40 (2.2)	1.79 (0.01)	94.8 (3.1)	95.0 (2.8)	16.8 (1.8)	15.4 (1.4)†

†Pre- and Post-exercise body fat measures were significantly different ($p < 0.0108$)

Both exercise groups exhibited improvements in measures of cardiorespiratory capacity, while CON subject scores were not significantly different (Table 6.3). Though both exercise groups experienced statistically significant improvements in cardiorespiratory capacity, relative pre- to post-exercise increases recorded in FIT subjects were approximately 3% greater than those of MOV subjects.

Table 6.3. Mean (SEM) total treadmill time and predicted $V0_2$ max of control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects before (Pre) and following (Post) the 12-week intervention.

Group	Time	Treadmill Time, seconds	$V0_2$ Max, mL/kg/min
CON	Pre	663 (24.3)	39.4 (1.33)
	Post	640 (21.7)	38.4 (1.18)
	p-value	0.1071	0.0996
FIT	Pre	665 (30.3)	38.8 (1.63)
	Post	749 (25.2)	42.9 (1.45)
	p-value	< 0.0001	< 0.0001
MOV	Pre	640 (26.2)	38.5 (1.35)
	Post	703 (25.5)	41.4 (1.27)
	p-value	< 0.0001	< 0.0001

Endurance of the trunk musculature remained unchanged in CON subjects over the study duration ($p < 0.1719$), but a number of improvements were exhibited in subjects who exercised. Increases in trunk flexion and extension endurance test times were statistically significant in both exercise groups ($p < 0.0001$); however, increases in trunk flexion (75%, 60-second increase) and extension (65%, 48-second increase) endurance test times for FIT subjects were somewhat greater than the 55% (48-second increase in flexion) and 42% (37-second increase in extension) average improvements experienced by MOV subjects. Whereas no pre-to post-training differences were exhibited in lateral bend endurance test times of MOV subjects ($p > 0.2067$), left-side times were 42% (23-second increase) greater in FIT subjects following training ($p = 0.0040$). Right-side lateral bend endurance times in all subjects did not change over the course of the study ($p = 0.8453$).

In all subjects, the maximum number of push-ups that could be performed was significantly greater at the end of the study. However, average improvements of 62 % (27 push-ups) experienced by FIT subjects ($p < 0.0001$) exceeded what was seen in MOV (38%, 14 push-ups) ($p < 0.0001$) and CON subjects (11%, 5 push-ups) ($p = 0.0353$).

Upper body power output measures significantly increased for FIT subjects at all but one testing load (50 lb) and for MOV subjects at all but two testing loads (30 and 50 lb). However, at the 70 and 90 lb testing loads, CON subjects also exhibited improvements in upper body power (Table 6.4).

Results from vertical jump and Keiser® squat machine tests indicated that members of both exercise groups experienced significant gains in lower body power output. Similar improvements in vertical jump heights of approximately 5% (≈ 3 cm) were recorded in FIT ($p = 0.0002$) and MOV ($p = 0.0001$) subjects while no differences were detected in CON subjects ($p = 0.5119$). Keiser® squat machine tests also revealed increased lower body power output in FIT and MOV subjects (Table 6.5). At all but one testing load (40 lb), Keiser® squat machine power readings in CON subjects were unchanged over the study duration.

Table 6.4. Mean (SEM) peak upper body power output (in Watts) of control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects before (Pre) and following (Post) the 12-week intervention. Values were recorded from the Keiser® chest press machine display.

Group	Time	Keiser® Chest Press Machine Load Setting (lb)				
		30†	**50‡**	**70‡**	**90‡**	**110†**
CON	Pre	315 (9.7)	375 (12.1)	407 (16.5)	408 (17.6)	387 (20.2)
	Post	330 (11.8)	381 (15.2)	408 (16.4)	414 (20.3)	383 (22.3)
	p-value	0.0773	0.0675	0.0379	0.0015	0.8010
FIT	Pre	322 (12.6)	408 (20.7)	435 (30.4)	423 (28.5)	400 (30.9)
	Post	364 (17.4)	416 (15.1)	447 (20.4)	456 (23.5)	445 (25.8)
	p-value	< 0.0001	0.0675	0.0379	0.0015	0.0095
MOV	Pre	317 (11.2)	372 (17.6)	384 (22.2)	393 (24.8)	362 (28.7)
	Post	332 (14.7)	386 (18.9)	418 (24.5)	434 (34.0)	408 (38.3)
	p-value	0.0623	0.0675	0.0379	0.0015	0.0031

†Statistically significant group×time interaction effects were detected for both the 30 lb (p = 0.0392) and 110 lb (p = 0.0305) testing loads; p-values correspond to within-group comparisons based on the least-square means procedures.
‡No statistically significant group×time interaction effects were detected for the 50 lb (p = 0.7790), 70 lb (p = 0.1535), and 90 lb (p = 0.1679) testing loads; p-values in the table correspond to the ANOVA within-subject ("time") factor.

In all groups (including CON subjects), grip strength measures were significantly greater at the end of the study than they were at the beginning. Grip strength measures increased in all subjects by an average of 4% (2 kg increase) for both the left (p < 0.0001) and right (p = 0.0012) hands.

Sit-and-reach measurements were unchanged in CON (p = 0.1333) and FIT (p = 0.8346) subjects, but MOV subjects were able to reach approximately 22% (≈ 4 cm) further after completing the exercise program (p < 0.0001).

Table 6.5. Mean (SEM) peak lower body power output (in Watts) of control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects before (Pre) and following (Post) the 12-week intervention. Values were recorded from the Keiser® squat machine display.

Group	Time	Keiser® Squat Machine Load Setting (lb)				
		40	60	90	120	150
CON	Pre	187 (10.2)	347 (16.3)	602 (24.0)	832 (28.6)	1064 (34.6)
	Post	219 (12.0)	376 (17.1)	617 (18.6)	861 (27.5)	1082 (31.0)
	p-value	0.0288	0.1070	0.4836	0.1919	0.5069
FIT	Pre	196 (12.9)	374 (17.7)	619 (21.6)	865 (27.5)	1090 (30.8)
	Post	301 (13.1)	470 (18.3)	745 (24.7)	1035 (29.0)	1269 (35.6)
	p-value	< 0.0001	< 0.0001	< 0.0001	< 0.0001	< 0.0001
MOV	Pre	199 (11.2)	359 (18.0)	614 (29.1)	827 (33.9)	1045 (44.8)
	Post	250 (15.0)	425 (19.2)	692 (29.3)	952 (32.9)	1170 (41.4)
	p-value	0.0002	0.0002	0.0003	< 0.0001	< 0.0001

Functional Movement Screen™ Scores

When the data were pooled, FMS task scores in the FIT and MOV subjects were not different between the pre- and post-exercise testing sessions; however, the shoulder mobility scores of CON subjects increased over the course of the study (Table 6.6). Upon closer inspection of the individual datasets it was revealed that FMS scores were somewhat variable (Table 6.7), making it difficult to ascertain if exercise resulted in consistent changes in general movement behaviours. Interpretation was particularly complicated by the finding that CON subject FMS scores were not stable over the study duration, as intra-class correlation coefficients ranged from 0.2713 (FMS Push-Up) to 0.6106 (FMS Active Straight-Leg Raise) in the individual FMS task scores of CON subjects.

Table 6.6. Mean (SEM) scores of Functional Movement Screen (FMS) tasks performed at the beginning (Pre) and end (Post) of the study. Data from control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects are included.

FMS Task†	CON			FIT			MOV		
	Pre	Post	*p*-value‡	Pre	Post	*p*-value‡	Pre	Post	*p*-value‡
ASLR	1.8 (0.17)	2.0 (0.16)	0.3984	1.6 (0.14)	1.6 (0.14)	1.000	1.8 (0.14)	1.9 (0.15)	0.6875
HSTP	1.9 (0.08)	1.9 (0.09)	1.000	2.0 (0.00)	2.0 (0.00)	1.000	2.0 (0.05)	1.8 (0.11)	0.5000
ILNG	2.1 (0.13)	2.3 (0.09)	0.7500	2.4 (0.11)	2.5 (0.12)	0.6250	2.1 (0.15)	2.3 (0.17)	0.4375
PSHP	2.0 (0.19)	1.8 (0.12)	0.2168	1.8 (0.20)	1.7 (0.17)	0.6699	1.9 (0.20)	1.7 (0.17)	0.3984
RTRY	2.0 (0.17)	1.8 (0.14)	0.4839	2.0 (0.15)	2.3 (0.13)	0.1719	1.8 (0.14)	1.9 (0.15)	0.5625
SHLD	1.7 (0.18)	2.2 (0.13)	0.0479	1.6 (0.17)	1.8 (0.19)	0.3594	1.8 (0.19)	1.8 (0.20)	1.000
DSQT	1.3 (0.13)	1.2 (0.08)	0.5313	1.4 (0.16)	1.1 (0.07)	0.1250	1.4 (0.14)	1.2 (0.17)	0.3750
COMP	12.5 (0.54)	12.8 (0.46)	0.6619	12.8 (0.41)	13.1 (0.44)	0.3962	12.8 (0.62)	12.6 (0.52)	0.8883

†ASLR = Active Straight-Leg Raise; HSTP = Hurdle Step; ILNG = In-Line Lunge; PSHP = Push-Up; RTRY = Rotary Stability; SHLD = Shoulder Mobility; DSQT = Deep Squat; COMP = Composite (Total) FMS score
‡Wilcoxon signed-rank tests were used to make within-group (Pre-Post) comparisons in FMS scores.

Table 6.7. Number of subjects whose FMS scores increased (↑), decreased (↓), or did not change (—) over the study duration. Data from control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects are included.

FMS Task†	CON			FIT			MOV		
	↑	↓	—	↑	↓	—	↑	↓	—
ASLR	4	2	10	5	6	7	4	2	14
HSTP	1	2	13	0	0	18	0	2	18
ILNG	1	1	14	3	1	14	5	2	13
PSHP	1	6	9	6	4	8	3	5	12
RTRY	3	6	7	6	2	10	5	2	13
SHLD	9	3	4	5	2	11	5	2	13
DSQT	1	3	12	1	6	11	1	3	16
COMP	8	6	2	9	5	4	8	5	7

†ASLR = Active Straight-Leg Raise; HSTP = Hurdle Step; ILNG = In-Line Lunge; PSHP = Push-Up; RTRY = Rotary Stability; SHLD = Shoulder Mobility; DSQT = Deep Squat; COMP = Composite (Total) FMS score

Low-Back Loading Demands in Simulated Firefighting Tasks

Although some differences in peak L4/L5 joint compression and reaction shear forces were detected between the pre- and post-training biomechanical testing sessions, there was no clear indication that fitness- or movement-centric exercise induced lasting spine “load-sparing” adaptations in movement behaviour. In only one task (Hose Pull) was the pre- to post-exercise peak L4/L5 joint compressive loading response of FIT and MOV subjects different from that of the CON subjects (Table 6.7). After the 12-week exercise intervention, peak L4/L5 compression forces during the Hose Pull task were not different in CON subjects (p = 0.9684), whereas MOV subjects imposed lower magnitude peak L4/L5 compression forces (p = 0.0009) and FIT subjects imposed higher peak magnitude L4/L5 compression forces (p = 0.0043). In the other tasks where pre- to post-exercise differences in peak L4/L5 compression forces were detected (Symmetrical Lift, Unilateral Push, Ceiling Pull), MOV, FIT, and CON subjects all responded similarly (Table 6.8).

In only two tasks (Asymmetrical Lift, Symmetrical Lift) were the pre- to post-exercise responses in peak L4/L5 reaction shear force magnitudes different between CON, FIT, and MOV subjects (Tables 6.8 and 6.9). When performing Asymmetrical Lifts, MOV subjects experienced lower magnitude peak L4/L5 A/P reaction shear forces in post-exercise testing than in pre-exercise testing (p = 0.0223), whereas no pre- to post-exercise differences in peak L4/L5 A/P reaction shear forces were noted in CON (p = 0.0681) or FIT (p = 0.5909) subjects (Table 6.9). In contrast, no pre- to post-exercise differences in peak L4/L5 M/L reaction shear forces were detected in FIT (p = 0.2485) or MOV (p = 0.0598) subjects when performing Symmetric Lifts, but post-exercise peak L4/L5 M/L reaction shear force magnitudes were greater in CON subjects than they were in pre-exercise testing (p = 0.0276) (Table 6.10). In other tasks where pre- to post-exercise differences in peak L4/L5 reaction shear forces were detected (Ceiling Breach, Ceiling Pull), MOV, FIT, and CON subjects all responded similarly (Tables 6.9 and 6.10).

Table 6.8. Mean (SEM) peak L4/L5 compression forces (kN) quantified during the performance of laboratory-simulated tasks performed at the beginning (Pre) and end (Post) of the study. Data are reported for control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects.

Task†	**CON**		**FIT**		**MOV**		***p*-value**‡		
	Pre	**Post**	**Pre**	**Post**	**Pre**	**Post**	**group**	**time**	**group×time**
ASYM	7.57 (0.38)	7.58 (0.42)	7.53 (0.26)	7.60 (0.30)	7.55 (0.31)	7.54 (0.30)	0.9976	0.8795	0.9687
SYMM	6.95 (0.33)	7.21 (0.32)	7.34 (0.21)	7.61 (0.29)	7.25 (0.29)	7.45 (0.29)	0.5983	0.0408	0.9592
PUSH	2.06 (0.10)	2.26 (0.13)	2.23 (0.10)	2.40 (0.17)	2.04 (0.13)	2.28 (0.15)	0.5832	0.0039	0.9110
PULL	3.22 (0.18)	3.49 (0.32)	3.03 (0.14)	3.29 (0.20)	3.16 (0.21)	3.14 (0.21)	0.7505	0.1338	0.4674
CBRC	3.70 (0.38)	3.79 (0.42)	4.32 (0.40)	4.30 (0.28)	4.81 (0.34)	5.11 (0.32)	0.0234	0.7318	0.6063
CPUL	2.83 (0.29)	2.57 (0.24)	3.14 (0.23)	2.75 (0.17)	3.09 (0.20)	3.00 (0.23)	0.4602	0.0054	0.3705
FENT	12.8 (1.15)	12.9 (1.21)	12.8 (0.94)	13.5 (0.90)	12.6 (1.03)	13.1 (1.06)	0.9684	0.1176	0.7322
CHOP	7.65 (0.61)	7.43 (0.49)	7.76 (0.39)	7.80 (0.40)	7.13 (0.42)	7.58 (0.44)	0.7730	0.6118	0.3116
HPUL	5.10 (0.34)	5.09 (0.41)	4.34 (0.23)	4.99 (0.35)	5.09 (0.47)	4.66 (0.34)	—	—	0.0030

†ASYM = Asymmetrical Lift; SYMM = Symmetrical Lift; PUSH = Unilateral Push; PULL = Unilateral Pull; CBRC = Ceiling Breach; CPUL = Ceiling Pull; FENT = Forcible Entry; CHOP = Overhead Chop; HPUL = Hose Pull
‡General linear model ANOVAs with one between-subject factor (group: CON vs. FIT vs. MOV) and one within-subject factor (time: Pre vs. Post) were performed to examine the impact of exercise on peak L4/L5 compression forces during task execution.

Table 6.9. Mean (SEM) peak L4/L5 anterior/posterior reaction shear forces (N) quantified during the performance of laboratory-simulated tasks performed at the beginning (Pre) and end (Post) of the study. Data are reported for control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects.

Task†	**CON**		**FIT**		**MOV**		***p*-value**‡		
	Pre	**Post**	**Pre**	**Post**	**Pre**	**Post**	**group**	**time**	**group×time**
ASYM	371 (36.9)	420 (39.3)	322 (29.0)	335 (31.4)	371 (24.0)	318 (25.0)	—	—	0.0140
SYMM	339 (32.3)	358 (35.3)	291 (29.6)	288 (25.7)	311 (24.9)	310 (34.6)	0.3318	0.7543	0.8435
PUSH	112 (6.4)	112 (7.0)	119 (10.7)	115 (8.6)	129 (11.5)	118 (9.4)	0.6084	0.3757	0.6929
PULL	227 (12.1)	243 (12.4)	232 (17.3)	224 (10.6)	230 (13.5)	224 (11.3)	0.8547	0.9298	0.4094
CBRC	195 (13.8)	188 (19.1)	218 (18.5)	199 (15.6)	209 (17.3)	230 (26.9)	0.3483	0.6611	0.2936
CPUL	222 (16.6)	204 (10.6)	235 (13.0)	212 (9.1)	246 (16.5)	232 (17.0)	0.3333	0.0354	0.8825
FENT	232 (19.9)	227 (31.6)	227 (19.9)	239 (24.0)	269 (23.1)	240 (22.7)	0.6741	0.4347	0.2349
CHOP	731 (46.2)	745 (38.8)	711 (36.4)	738 (37.7)	749 (38.7)	743 (28.5)	0.8974	0.5773	0.7879
HPUL	325 (23.6)	296 (26.3)	264 (23.5)	248 (20.9)	261 (23.2)	241 (17.3)	0.0901	0.0679	0.9083

†ASYM = Asymmetrical Lift; SYMM = Symmetrical Lift; PUSH = Unilateral Push; PULL = Unilateral Pull; CBRC = Ceiling Breach; CPUL = Ceiling Pull; FENT = Forcible Entry; CHOP = Overhead Chop; HPUL = Hose Pull

‡General linear model ANOVAs with one between-subject factor (group: CON vs. FIT vs. MOV) and one within-subject factor (time: Pre vs. Post) were performed to examine the impact of exercise on peak L4/L5 A/P reaction shear forces during task execution.

Table 6.10. Mean (SEM) peak L4/L5 medial/lateral reaction shear forces (N) quantified during the performance of laboratory-simulated tasks performed at the beginning (Pre) and end (Post) of the study. Data are reported for control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects.

Task†	CON		FIT		MOV		*p*-value‡		
	Pre	**Post**	**Pre**	**Post**	**Pre**	**Post**	**group**	**time**	**group×time**
ASYM	125 (11.2)	147 (14.7)	96 (4.8)	94 (3.9)	100 (5.0)	99 (6.9)	0.0003	0.1842	0.0816
SYMM	76 (6.3)	88 (8.9)	83 (4.5)	77 (5.1)	82 (5.6)	92 (5.8)	—	—	0.0343
PUSH	87 (8.7)	80 (8.9)	74 (5.3)	73 (5.8)	67 (6.1)	62 (5.1)	0.0824	0.2438	0.7614
PULL	152 (15.3)	146 (13.4)	121 (8.4)	113 (8.1)	117 (9.9)	111 (10.5)	0.0492	0.0735	0.9887
CBRC	128 (8.7)	132 (15.1)	160 (10.5)	121 (8.0)	140 (14.6)	129 (11.4)	0.5723	0.0051	0.1021
CPUL	147 (11.4)	130 (9.1)	168 (15.1)	127 (9.0)	133 (7.7)	117 (8.1)	0.1625	0.0007	0.2216
FENT	243 (13.1)	243 (15.6)	249 (10.0)	231 (12.2)	237 (13.2)	221 (13.6)	0.6648	0.1008	0.5193
CHOP	191 (12.6)	191 (11.8)	183 (9.9)	178 (10.8)	190 (11.5)	195 (13.0)	0.6724	0.9798	0.7442
HPUL	174 (11.0)	161 (10.8)	155 (9.9)	155 (12.1)	168 (11.0)	150 (12.3)	0.7044	0.0605	0.3557

†ASYM = Asymmetrical Lift; SYMM = Symmetrical Lift; PUSH = Unilateral Push; PULL = Unilateral Pull; CBRC = Ceiling Breach; CPUL = Ceiling Pull; FENT = Forcible Entry; CHOP = Overhead Chop; HPUL = Hose Pull
‡General linear model ANOVAs with one between-subject factor (group: CON vs. FIT vs. MOV) and one within-subject factor (time: Pre vs. Post) were performed to examine the impact of exercise on peak L4/L5 M/L reaction shear forces during task execution.

In Tables 6.11 and 6.12, it can be seen that when the peak L4/L5 joint compression force was imposed during task performance, there were some minor differences in the lumbar spine orientation between the pre- and post-exercise testing sessions. However, with only one exception, pre- to post-exercise differences were consistent between MOV, FIT, and CON subjects. When performing the Unilateral Push task, the lumbar spine was more deviated about the axial twist axis in FIT subjects following exercise (p = 0.0365) and less deviated following exercise in MOV subjects (p = 0.0141). No pre- to post-exercise differences were detected in this variable in CON subjects (p = 0.2448).

Table 6.11. Absolute value of the lumbar spine angle about the lateral bend (Bend), axial twist (Twist), and flexion/extension (Flex/Ext) axes when the peak L4/L5 compression force was imposed. Mean (SEM) values calculated during simulated manual handling tasks performed at the beginning (Pre) and end (Post) of the study are reported for control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects.

Task[†]	**Axis**	**CON**		**FIT**		**MOV**		***p*-value**[‡]		
		Pre	**Post**	**Pre**	**Post**	**Pre**	**Post**	**group**	**time**	**group×time**
ASYM	Bend	7.2 (1.2)	6.1 (1.1)	6.5 (0.9)	5.8 (0.8)	4.8 (0.6)	4.3 (0.8)	0.1214	0.1636	0.8964
	Twist	3.7 (0.7)	3.1 (0.7)	2.8 (0.4)	2.0 (0.3)	3.2 (0.6)	2.8 (0.6)	0.3892	0.0675	0.8782
	Flex/Ext	57.4 (2.7)	56.3 (2.7)	56.1 (1.7)	56.3 (1.8)	54.0 (1.7)	52.4 (1.8)	0.3347	0.3164	0.6450
SYMM	Bend	2.3 (0.3)	1.6 (0.3)	1.5 (0.3)	2.4 (0.5)	1.9 (0.3)	2.1 (0.3)	0.9636	0.6802	0.0741
	Twist	3.2 (0.5)	2.3 (0.4)	1.9 (0.4)	1.3 (0.2)	2.5 (0.4)	1.9 (0.3)	0.0424	0.0091	0.9355
	Flex/Ext	48.8 (2.6)	51.8 (3.2)	53.3 (2.3)	52.1 (2.5)	49.6 (2.1)	47.0 (2.2)	0.3755	0.7701	0.1026
PUSH	Bend	3.0 (0.5)	3.4 (0.6)	2.7 (0.6)	2.5 (0.6)	3.4 (0.6)	2.6 (0.5)	0.6833	0.5526	0.4180
	Twist	3.5 (0.9)	2.7 (0.6)	2.1 (0.5)	3.5 (0.5)	4.1 (0.7)	2.6 (0.5)	—	—	0.0055
	Flex/Ext	3.7 (0.7)	5.9 (1.0)	5.7 (1.1)	7.9 (2.0)	4.9 (0.8)	6.0 (1.0)	0.3820	0.0097	0.7375
PULL	Bend	4.5 (0.6)	5.0 (1.2)	4.5 (0.7)	4.1 (0.8)	3.7 (0.6)	3.6 (0.7)	0.4953	0.9895	0.7344
	Twist	25.1 (1.6)	24.5 (1.6)	21.2 (1.0)	18.2 (1.4)	19.6 (1.6)	17.5 (1.1)	0.0009	0.0511	0.5881
	Flex/Ext	13.3 (2.4)	12.2 (2.4)	16.0 (2.1)	18.2 (2.1)	15.4 (1.8)	15.4 (2.0)	0.3264	0.6915	0.3392

[†]ASYM = Asymmetrical Lift; SYMM = Symmetrical Lift; PUSH = Unilateral Push; PULL = Unilateral Pull; CBRC = Ceiling Breach; CPUL = Ceiling Pull; FENT = Forcible Entry; CHOP = Overhead Chop; HPUL = Hose Pull
[‡]General linear model ANOVAs with one between-subject factor (group: CON vs. FIT vs. MOV) and one within-subject factor (time: Pre vs. Post) were performed to examine the impact of exercise on lumbar spine deviation during task execution.

Table 6.12. Absolute value of the lumbar spine angle about the lateral bend (Bend), axial twist (Twist), and flexion/extension (Flex/Ext) axes when the peak L4/L5 compression force was imposed. Mean (SEM) values calculated during simulated fireground tasks performed at the beginning (Pre) and end (Post) of the study are reported for control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV) subjects.

Task†	Axis	CON		FIT		MOV		*p*-value‡		
		Pre	Post	Pre	Post	Pre	Post	group	time	group×time
CBRC	Bend	7.5 (1.2)	6.4 (1.4)	9.0 (1.1)	7.4 (1.2)	7.0 (1.0)	6.1 (1.0)	0.4280	0.0870	0.8950
	Twist	4.5 (1.0)	4.1 (0.7)	4.6 (0.9)	3.8 (0.9)	3.6 (0.6)	3.4 (0.7)	0.6642	0.3235	0.8506
	Flex/Ext	10.2 (1.7)	11.5 (1.6)	12.4 (1.1)	13.4 (1.5)	12.3 (1.4)	15.9 (1.2)	0.1847	0.0176	0.3212
CPUL	Bend	7.1 (1.0)	7.0 (1.0)	4.9 (0.8)	4.8 (0.8)	4.8 (0.7)	6.7 (0.9)	0.1307	0.3432	0.2692
	Twist	3.6 (0.8)	3.4 (0.6)	4.6 (0.8)	4.6 (0.9)	4.7 (0.9)	5.6 (0.9)	0.1435	0.7864	0.8083
	Flex/Ext	23.6 (2.2)	23.0 (2.4)	23.3 (1.7)	25.0 (1.7)	25.9 (2.1)	25.8 (1.7)	0.5737	0.7020	0.6027
FENT	Bend	13.8 (1.6)	12.3 (1.7)	12.2 (1.4)	11.2 (1.3)	11.6 (0.9)	10.9 (0.8)	0.5495	0.0381	0.8411
	Twist	17.1 (1.8)	17.3 (2.3)	15.5 (1.7)	13.6 (1.4)	14.8 (1.8)	12.8 (1.2)	0.2869	0.1605	0.4977
	Flex/Ext	22.0 (1.6)	25.3 (1.8)	19.4 (2.0)	24.5 (1.7)	21.8 (1.7)	25.7 (1.5)	0.6943	< 0.0001	0.5695
CHOP	Bend	4.0 (0.9)	3.8 (0.8)	4.9 (0.9)	3.8 (0.7)	5.2 (1.0)	4.1 (1.0)	0.8497	0.0069	0.3252
	Twist	5.3 (1.5)	4.5 (0.7)	4.7 (0.9)	3.9 (0.8)	6.4 (1.1)	5.0 (1.1)	0.4826	0.1573	0.9222
	Flex/Ext	54.8 (3.3)	57.8 (2.8)	55.0 (1.9)	55.5 (2.5)	48.4 (2.4)	52.2 (1.9)	0.1476	0.0137	0.3229
HPUL	Bend	5.7 (0.9)	4.9 (0.9)	4.3 (1.0)	6.3 (1.1)	4.8 (0.9)	5.2 (0.9)	0.9490	0.3835	0.1445
	Twist	7.8 (1.6)	8.9 (1.4)	8.7 (1.7)	10.0 (1.4)	10.6 (1.7)	6.8 (1.0)	0.8256	0.6546	0.0717
	Flex/Ext	39.1 (2.9)	40.9 (2.4)	39.5 (1.9)	37.4 (1.5)	37.6 (1.7)	36.8 (1.6)	0.5619	0.6918	0.2358

†ASYM = Asymmetrical Lift; SYMM = Symmetrical Lift; PUSH = Unilateral Push; PULL = Unilateral Pull; CBRC = Ceiling Breach; CPUL = Ceiling Pull; FENT = Forcible Entry; CHOP = Overhead Chop; HPUL = Hose Pull
‡General linear model ANOVAs with one between-subject factor (group: CON vs. FIT vs. MOV) and one within-subject factor (time: Pre vs. Post) were performed to examine the impact of exercise on lumbar spine deviation during task execution.

6.4. Discussion

Examined in this study was the notion that in comparison to a fitness-centric exercise approach (i.e., exercise designed primarily to improve measures of physical fitness), movement-centric exercise (i.e., exercise hypothesized to alter habitual patterns of coordination and control) would constitute a preferred exercise-based low-back injury prevention strategy for firefighters. Specifically, it was hypothesized that firefighters who completed movement-centric exercise would elect to perform simulated job tasks in ways that attenuated peak low-back loading (via movement behaviour modifications). Although both movement- and fitness-centric exercise programs resulted in improvements in physical fitness (i.e., body composition, cardiorespiratory capacity, muscular strength, power, endurance, and flexibility), it could not be concluded that that either 12-week exercise program produced consistent in peak low-back loading responses to simulated job demands.

The exercise programs were intended to induce two different types of general adaptations, namely fitness- and movement-related adaptations. Besides being unlikely that such adaptations can be considered mutually exclusive and given that subjects were justifiably unwilling to permit decreases in physical fitness given the hazardous nature of their work, it was not possible to design exercise programs that could be clearly differentiated based on desired outcomes *a priori*. FIT and MOV program design considerations were based on hypotheses derived from previous experiences and principles of exercise science, resulting in programs that varied not only with respect to the focus of coaching instruction and feedback, but also with respect to exercises included, progressions followed, volumes, intensities, etc.. Consequently, a battery of physical fitness tests and movement screening tasks were included to evaluate *a posteriori* if desired exercise adaptations were elicited by the training programs. Results clearly indicated that exercise improved physical fitness and the improvements experienced by FIT and MOV subjects were generally of similar magnitudes. This was somewhat unexpected given that

FIT subjects were essentially "training for the fitness tests" by being directed to focus their attention on performance outcomes during training (external focus), whereas MOV subjects were directed to focus on the *way* they performed exercises rather than on the impact that their movements had on performance outcomes (internal focus). Recent research has consistently demonstrated that individuals whose focus of attention is directed externally achieve superior and more rapid improvements in performance outcomes than when instructed to focus internally (Wulf et al. 2010). However, because MOV subjects were required to keep a detailed training log (information used to progress exercise), they were not strictly blinded to their performance outcomes and were likely motivated to better previous performance results despite being directed to focus their attention to their movements. This lack of experimental control could explain why FIT and MOV subjects experienced similar improvements physical fitness, but was not considered to be a study limitation given that it is likely impractical and undesirable to conceal performance outcomes in real-world settings. Moreover, movement education alone has been shown to have less impact on musculoskeletal loading and injury risk measures than does a combined approach incorporating both strength training and movement education (Herman et al. 2009). Nevertheless, it is important to highlight that equal improvements were not exhibited in all measures of physical fitness (i.e., push-ups, trunk muscle endurance, and sit-and-reach). This was probably a reflection of between-program differences in the amount of time and effort spent executing these exercises (or related supplemental movements) over the course of the study (i.e., due to specificity of training).

It could be concluded from the FMS results that the MOV program was not effective in inducing general movement-related exercise adaptations. From a motor learning perspective, movements are "learned" when desired/intended changes are retained (beyond training) and transfer to other related, yet unpracticed activities. Learning can be affected by the frequency, timing, and/or type of feedback provided in addition to the organization and structure of training (Wulf et al. 2010), and it is certainly possible that such factors were inaptly incorporated in the MOV program. Though the study design limited the ability to directly address such a question, day-to-day observations made by the MOV coach and the comparison of pre- and post-exercise FMS scores were used to yield some insight regarding MOV program effectiveness. However, any within and between exercise session movement-related improvements witnessed by the MOV coach could have been indicative of transient performance effects rather than genuine motor learning (Newell 2003), and it is also possible that the FMS constituted a poor transfer test with respect to study objectives. One salient limitation of using the FMS for the purposes described here was previously uncovered in a separate and more detailed analysis of the FMS data collected (Frost et al. 2011). Though Shultz et al. (2011) indicated that the test-retest reliability of FMS scores is satisfactory if captured within a 7-day period (Krippendorff's alpha = 0.6161), Frost and colleagues (2011) found that the FMS scores of CON subjects in this study were extremely variable over the 12-week data collection period and consequently raised concerns about the ability of the FMS to detect exercise adaptations under the conditions examined. Previous 6- and 7-week intervention studies used the FMS to document exercise adaptations (Cowen 2010; Goss et al. 2009; Kiesel et al. 2009), but the applicability of their results are limited by the fact that no control groups/conditions were included, individual FMS task scores were not independently analyzed (intra-individual variation in between-day FMS task scores could be "masked" in the composite score), and the parametric statistical tests employed may have led to false or misleading conclusions because FMS tasks are graded on an ordinal scale. Accordingly, given these questions regarding the test-retest reliability of the FMS, the finding that FMS scores were not influenced by either program must be interpreted cautiously. Future attempts to use the FMS in intervention research should be preceded by a more thorough and controlled examination of its test-retest reliability, especially since the intra-class correlation coefficients calculated for individual FMS tasks in this study ranged from less than 0.3 (low reliability) to less than 0.7 (moderate reliability) in subjects who did not exercise.

The peak low-back loading response to simulated task demands was apparently unaltered after 12 weeks of exercise. Any pre- to post-exercise changes in peak L4/L5 joint compression and reaction shear forces exhibited by MOV or FIT subjects were either consistent with those seen in CON subjects or were of biomechanically trivial magnitudes (i.e., exercise did not cause peak low-back loading levels to fall below or rise above recommended action limits for joint compression (NIOSH 1981) or reaction shear (McGill et al. 1998) forces). However, there are a number of issues that must be considered before concluding that exercise had little or no impact on the low-back loading response to simulated task demands. First, consistent with what was found when examining the pre- and post-exercise FMS scores (Frost et al. 2011), the pooled low-back loading results did not accurately characterize the responses of all study subjects. There were CON, FIT, and MOV subjects who exhibited peak low-back forces of greater, lesser, and equal magnitude in post-exercise testing sessions than in pre-exercise sessions, resulting in no apparent impact of exercise when the low-back loading data were aggregated. Experimental factors (e.g., anatomical landmark identification errors) could have accounted for some of the variability in low-back loading responses, but a number of model-building computational procedures were used to attenuate the impact of between-day measurement inconsistencies (e.g., segment end-points and coordinate systems were determined “functionally”) and previous research has examined the impact of exercise on various joint kinematic and kinetic measures using comparable methodology (Cochrane et al. 2010; Herman et al. 2009; Meyer et al. 2007; Willy and Davis 2011). Perhaps a better explanation for the inconsistent low-back loading findings is that whole-body movement strategies (and corresponding internal low-back loading responses) were found to be variable between individuals and testing days. Considerable inter- and intra-individual variability is an inherent characteristic of human movement (Newell and Corocos 1993); this can be attributed to the fact that the musculoskeletal linkage is endowed with numerous biomechanical degrees-of-freedom and thus motor task objectives can be satisfied using many different patterns of coordination and control (Newell and Corocos 1993;

Newell and Slifkin 1998). Though there is some evidence of commonality in the underlying movement patterns of skilled performers (e.g., Zanone and Kelso 1992), even experts with years of practice (e.g., elite athletes) do not typically exhibit invariant kinematic and kinetic patterns during repeated task execution (Bartlett et al. 2007b). In fact, it has been reported that experienced individuals often exhibit greater movement variability than do novices (Schorer et al. 2007; Wilson et al. 2008), possibly reflecting that previous experience and practice makes available more movement solutions to permit stable performance outcomes in complex environments (Handford et al. 1997; Riley and Turvey 2002). Against this backdrop, future studies incorporating more sophisticated methods of motion analyses (e.g., Daffertshofer et al. 2004; Graham et al. 2011; Hamill et al. 1999; Kulić et al. 2009) in single-subject experimental designs might be better suited to expose and monitor movement-related exercise adaptations (Bates 1996), especially in cases where data aggregation results in an average response that is different from those of the individual subjects (Dufek et al. 1995).

Since low-back loading estimates were based on solely measures of net L4/L5 joint reaction forces (shear) or moments (compression) without explicitly taking trunk muscle activation or lumbar posture into account (c.f., Cholewicki and McGill 1996), it is important to acknowledge that exercise adaptations could have been undetected or misrepresented in the low-back loading estimates. It is conceivable, for example, that some subjects were able to produce equivalent low-back moments using less trunk muscle (co-)activation following exercise (due increased strength and altered coordination). If so, these subjects could have experienced lower levels of low-back compression following exercise when executing tasks with fixed external moment demands (e.g., lifting), but the polynomial method of load estimation would have been insensitive to such an effect. Or, when performing tasks without fixed external moment demands (e.g., forcible entry), some subjects may have generated higher-magnitude low-back moments following exercise using relatively unchanged levels of trunk muscle (co-)activation (again due to increased strength and altered coordination). In this case, the polynomial might have overestimated the low-back compressive load penalties associated with producing higher magnitude low-back net moments following exercise. Clearly, a more sophisticated musculoskeletal modeling approach would be needed to appropriately address these questions (e.g., Cholewicki and McGill 1996). However, it was decided to use the polynomial method because it has been shown, on average, to produce peak low-back compression estimates that are not significantly different from those derived from more complex EMG- and optimization-assisted spine models (Gagnon et al. 2001). However, it is acknowledged that individual low-back loading responses may be poorly predicted based on the polynomial method (see Figure 7.1 in Chapter 7).

Study limitations notwithstanding, it remains tempting to conclude that this studied yielded largely negative or inconclusive findings, especially given that exercise-induced improvements in physical fitness were expected. However, it is important to emphasize that improvements in physical fitness can enhance on-the-job performance and prevent cardiorespiratory failure in firefighters, and thus components of the exercise programs tested in this study could be incorporated into existing initiatives that aim to attain and maintain firefighter health, wellness, and work ability (WFI 2008). In particular, it could be hypothesized that although the MOV program did not clearly or consistently alter the peak low-back loading response to simulated job demands, a movement-centric approach might help to prevent exercise-related musculoskeletal injuries, a problem commonly encountered by "occupational athletes" who use exercise as a tool to enhance and maintain work ability and reduce injury risk but often experience the opposite effects (Bylund and Björnstig 1999; Evans et al. 2005; Heir and Glomsaker 1996; Gruhn et al. 1999; Jones and Knapik 1999; Kaufman et al. 2000; Lauder et al. 2000; Loës and Jansson 2001; Poplin et al. 2011). It is also critical to highlight that work-related injury potential can also be impacted by factors not considered in this study (e.g., musculoskeletal load tolerance) but that are also influenced by exercise (e.g., tissue adaptations). Thus, firefighters who exercise could also alter their “margin of safety” at work without changing their habitual movement behaviours.

6.5. Conclusions

Tested in this study was the notion that exercise designed to prioritize movement-related exercise adaptations over physical fitness-related adaptations would alter how individuals elected to move and load their low-back when performing simulated work tasks. While the physical fitness of study subjects improved over the course of the 12-week exercise intervention, it could not be concluded that either exercise program consistently altered simulated occupational low-back loading demands. Firefighters who are physically fit are better able to perform essential job duties and avoid cardiorespiratory failure, but the results of this study suggest that improvements in physical fitness following 12 weeks of exercise may not necessarily reduce the low-back loading demands associated with firefighter task performance. However, more research is needed to better understand how individuals adapt to exercise, especially over a longer term, and what impact exercise adaptations have on movement behaviour, low-back loading, and hypothesized injury potential.

CHAPTER 7

General Discussion and Conclusions

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7.1. Global and Specific Thesis Contributions

Global Research Contributions

Given that fireground task and environmental characteristics cannot normally be controlled, it was argued in Chapter 2 that interventions designed to impact fire scene movement behaviour could form a basis for worker-focused low-back injury prevention efforts. Current firefighter low-back injury prevention efforts may be somewhat hampered by the fact that no previous attempts have been made to document low-back loading responses to “heavy”, “arduous”, “strenuous”, and “physically demanding“ to fireground-specific tasks. Data in this thesis suggest that fireground exposures could indeed be hazardous for low-back health (Chapters 3 and 6). In Chapter 3, it was found that despite considerable inter-subject variation in the low-back loading response to fixed external task and environmental constraints, opportunities to attenuate peak low-back loading demands through movement behaviour modification alone may be limited to a subset of fireground activities. Results of the study presented in Chapter 4 indicated that ankle mobility restrictions can alter movement behaviour and low-back injury potential when lifting. Motivated by results in Chapter 4, the study in Chapter 5 aimed to test the idea that a general whole-body movement screen, envisaged to uncover personal qualities hypothesized to promote potentially injurious movement behaviours, could be used to project the low-back loading response to lifting. Limited perhaps by the approach taken to interpret screen outcomes, no association was detected between screen scores and low-back loading demands during lifting. Adaptations induced by a proposed movement behaviour modification exercise program were compared to those induced from a more conventional firefighter exercise program in Chapter 6. Both exercise approaches improved physical fitness over the 12-week intervention study, but adaptations resulting from either approach did not lead to consistent changes in simulated occupational low-back loading demands and hypothesized injury potential. All told, low-back loading estimates included in this thesis were derived from 54 career firefighters who performed a battery of laboratory-simulated general manual material handling and fireground-specific tasks before and after a 12-week exercise intervention. Data from an additional 30 university student volunteers were incorporated in studying the impact of personal characteristics on the low-back loading response to simulated lifting or fireground tasks. Collectively, results of the original research presented in this thesis were based on over 5,500 exertions performed by a physically diverse group of test subjects who had a wide range of work experiences and who represented both men and women.

Specific Research Contributions

Study 1: Low-Back Loading Demands during Simulated Firefighting Tasks – Inter-Subject Variation and the Impact of Fatigue and Gender

In Chapter 3, a biomechanical model that was sensitive to inter- and intra-individual differences in movement strategies was used to quantify low-back loading during simulated firefighting tasks, and the impact of physical fatigue and gender on the peak low-back loading response was examined. Hypotheses were supported in that considerable inter-subject variation in peak low-back loading was noted and since low-back loading responses varied between genders and with fatigue. It was generally concluded that physical characteristics of individuals and tasks performed may influence peak low-back loading demands and injury potential in firefighters. Though low-back loading demands tended to decrease with fatigue, it may not necessarily be indicative of reduced injury potential given that lumbar spine system stability could decrease under fatiguing conditions. Despite considerable inter-subject variation noted, opportunities to attenuate fireground low-back loading demands via movement behaviour modification alone may be limited to a subset of activities. However, exploiting such opportunities could reduce cumulative loading exposures and perhaps preserve capacity to withstand loading demands associated with non-modifiable job duties.

Study 2: Ankle Immobilization alters Lifting Kinematics and Kinetics – Occupational Low-Back Loading Demands and Potential for Injury

In Chapter 4, the effect of unilateral ankle joint immobilization on the kinematics and kinetics of lifting was studied. Results supported the hypothesis that ankle immobilization, an artificially imposed functional impairment, would promote compensatory movement strategies and consequently alter the peak low-back loading response to lifting. When subjects adapted their preferred lifting strategy in response to the kinematic constraint, peak low-back reaction shear force magnitudes increased as a result. Moreover, when the ankle was immobilized, the lumbar spine tended to be more deviated at the instant when peak low-back compression force was imposed. Distal lower extremity injuries are commonly suffered by firefighters (Karter 2009), and functional impairments can persist long after acute treatment (e.g., ankle dorsiflexion restriction). Results provided justification for exploring the utility of whole-body movement screens in firefighter low-back injury prevention programs (Chapter 5), as identifying dysfunctional movement patterns could be used to devise personalized worker-focused interventions (e.g., joint mobilization treatment).

Study 3: FMS™ Scores and Occupational Low-Back Loading Demands – Whole-Body Movement Screening as an Ergonomic Tool?

In Chapter 5, peak low-back loading during lifting was compared between size-matched firefighters who scored greater or less than 14 on the Functional Movement Screen™ (FMS), a previously established musculoskeletal injury prediction cut-point. Results did not support the hypothesis that peak low-back loading demands would differ between high- and low-scorers, as no between-group differences were detected in peak low-back compression or reaction shear force magnitudes. Though it was concluded that scoring above or below 14 on the FMS does not project the low-back loading response to lifting, it was recognized that activity- or occupation-specific FMS tasks or different cut-points could conceivably be used for this purpose. Future attempts to modify or reinterpret FMS scoring are warranted given that several previous studies have revealed links between composite FMS scores and musculoskeletal complaints.

Study 4: Movement- vs. Fitness-Centric Exercise – Firefighter Fitness, Whole-Body Movement Qualities, and Occupational Low-Back Loading Outcomes

In Chapter 6, physical fitness measures, FMS scores, and low-back loading during job task simulations were compared between groups of firefighters who completed 12 weeks of movement- or fitness-centric exercise. Both exercise programs improved physical fitness, but FMS scores and simulated occupational low-back loading demands were not consistently impacted across individuals. Aggregated results did not support the hypothesis that between-group differences in exercise outcomes could be used to justify one exercise-based low-back injury prevention approach over the other. Improvements in physical fitness can enhance performance capabilities and prevent cardiorespiratory events in firefighters, but it is not clear that 12-weeks of exercise would consistently alter occupational low-back loading demands in firefighters. Given variability in individual responses, the short study duration, and limited number and nature of tasks examined, more research incorporating alternative biomechanical and statistical analyses is needed to better understand how individuals adapt to chronic exercise exposure and what impact these adaptations have on occupational movement behaviours, low-back loading demands, and hypothesized injury potential.

7.2. Experimental Approaches and Assumptions

Laboratory Simulations

Laboratory-simulated tasks were based on observations and external force measurements made during the Candidate Physical Abilities Test (CPAT) and by consulting with fire department training officers. Clearly, it was not practical in a laboratory environment to replicate the chaotic and extreme environmental conditions under which firefighters must sometimes perform. Therefore, simulations did not account for the potential influence that many other mental (cognitive, emotional) or physical (environmental, physiological) stressors could have on firefighter movement strategies and internal low-back loading patterns at an actual fire scene. Davis et al. (2002) previously demonstrated that some individuals respond to the presence of simultaneous mental workplace stressors by imposing higher magnitude low-back loads during task execution (via increased trunk muscle co-activation) than they do when performing the same tasks without exposure to mental stressors. It is thus possible that some firefighters would elicit similar responses when exposed to mental stressors at a fire scene. Conceivably, this limitation could be addressed in future field-based studies (i.e., during realistic training exercises) by exploiting recent advances in portable motion capture, electromyography, and force measurement technologies.

Other ways in which low-back loading responses to the laboratory-simulated tasks could differ from those during genuine firefighter task performance relate to the fact that simulations may not have accurately represented the impact of personal protective equipment on low-back mechanics. First, expiratory resistance offered by a self-contained breathing apparatus (SCBA) could force abdominal muscles to contribute to ventilation, possibly affecting low-back loading patterns (McGill et al. 1995) and altering the capacity of the lumbar spine system to support and transmit applied loads (Wang and McGill 2008). Second, while a weighted vest was used to simulate the mass of on-body personal protective equipment, the mass was located and distributed in a way that may have overestimated the low-back mechanical demands in comparison to those imposed during real fireground operations. Specifically, the mass of the vest was applied to the distal (superior) torso of subjects with body-harness interface pressures likely of greatest magnitudes at the shoulders; however, most SCBA harnesses are comprised of a system of straps, belts, and suspension rods designed to distribute a more proximally- and posteriorly-located mass (air tanks) between the shoulders and pelvis. Thus, it could be hypothesized that in comparison to carrying a conventional SCBA, wearing the weighted vest may have led to an overestimation of the low-back loading levels by eliciting greater levels of trunk muscle activation to offset higher magnitude external low-back moments (when the torso was non-vertical) or to prevent spinal buckling (when the torso was upright). Again, this limitation could be addressed in future research incorporating different measurement tools (e.g., inertial motion tracking systems) so that biomechanical exposure data could be acquired while subjects are wearing authentic personal protective equipment.

Low-Back Loading Calculations

In Chapters 4, 5, and 6, a polynomial method developed by McGill et al. (1996b) was used to generate estimates of low-back compression loading during the performance of simulated firefighting tasks. Although a number of limitations of this method were already addressed throughout this thesis, it is important to consider what additional information might have been gained if a more sophisticated modeling approach had been used. The polynomial method produced low-back compression force estimates based only on the measured three-dimensional net L4/L5 joint moment of force. Thus, differences in low-back loading responses between subjects and over time may have gone undetected if due to inter- and intra-individual variations in lumbar spine kinematics and/or trunk muscle activation patterns. An anatomically detailed EMG-driven musculoskeletal model would have been required to detect such differences (e.g., Cholewicki and McGill 1996), but it was not possible to overcome equipment malfunctions (unrecoverable data) and experimental constraints encountered to permit the integration of EMG measurements into low-back load estimates. Given the nature of the lifting tasks examined in Chapters 4 and 5, it was decided that using the polynomial could be justified on the basis that low-back compressive load estimates during lifting are highly correlated with the magnitude of net joint moments of force (correlation coefficients > 0.9) (Dieën and Kingma 2005; McGill et al. 1996b). Not surprisingly as a result, peak low-back compression force estimates produced using the polynomial are not significantly different, on average, from those derived using EMG- and optimization-assisted musculoskeletal spine models during lifting exertions (Gagnon et al. 2001). Therefore, using the polynomial to calculate low-back compression forces during lifting was deemed adequate in Chapters 4 and 5.

There was some concern that using the polynomial to estimate compression forces during simulated firefighting tasks would be inappropriate because, as discussed previously, wearing the weighted vest may have required subjects to elicit higher-than-expected levels of trunk muscle co-activation in order to maintain stable lumbar spine system behaviour. The manual handling data used to develop the polynomial were collected from subjects who were not exposed to the additional burden of on-body load carriage (McGill et al. 1996b). Thus, it could not be assumed that the polynomial would yield representative estimates of low-back compression forces during the performance of firefighting tasks in Chapter 6. To test this assumption, a subset of the data in Chapter 3 was used to compute peak low-back compression forces using both an EMG-assisted musculoskeletal model (Cholewicki and McGill 1996) and the polynomial (McGill et al. 1996b). Only baseline (non-fatigued) data from male subjects in Chapter 3 were included in this ancillary analysis since the firefighters in Chapter 6 were also male and because they were permitted full recovery time between trials and tasks. Peak compression force estimates derived using both modeling approaches were compared. Although there were a number of discrepancies between the polynomial- and EMG-based peak compression estimates (Figure 7.1), a statistically significant linear relationship was detected between outputs from the approaches (r^2 = 0.6989; $p < 0.0001$). Hence, it could be argued that even if absolute values of the compression estimates would have been different if using a more sophisticated musculoskeletal model in Chapter 6, the polynomial was adequate for making relative (within-subject) comparisons if it was assumed that trunk muscle activation patterns or lumbar spine kinematics were uninfluenced by exercise. While lumbar spine motion was captured and analyzed in Chapter 6, EMG measurement system malfunctions prohibited the ability to maintain uninterrupted connections between the wireless EMG signal transmitters and receivers during data collection sessions.[1] As a consequence, most EMG data sets were incomplete and it was thus not possible to determine if trunk muscle activation patterns were different between the pre- and post-exercise testing sessions.

[1] EMG data packets were frequently lost due to interference from other wireless data transmission sources. The data collection facility was outfitted with dynamic data networks; it was therefore not possible to dedicate a unique EMG data transmission frequency without encountering transmission conflicts and delays caused by data that were simultaneously transmitted by other devices in the facility.

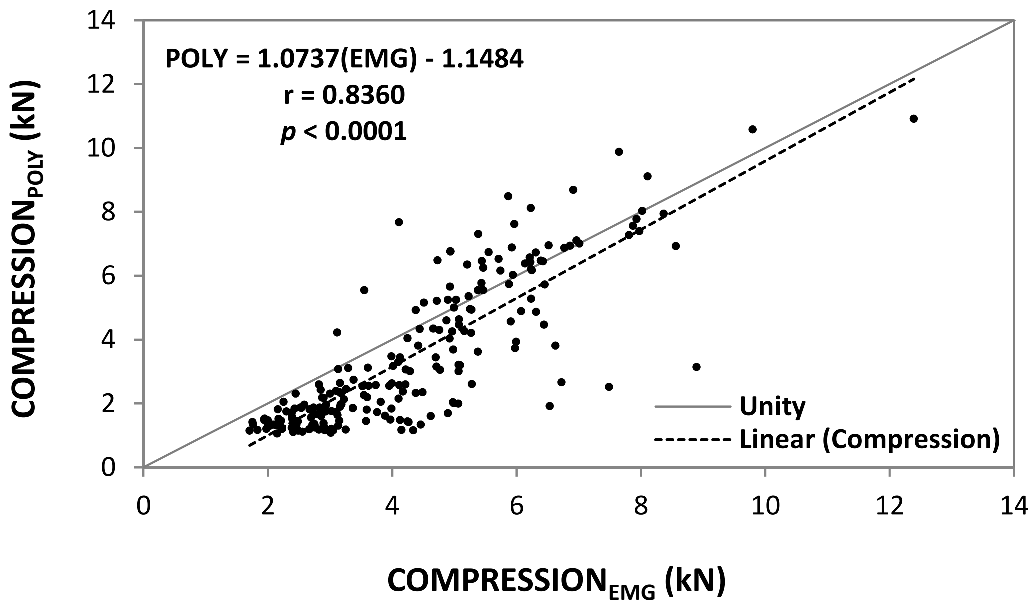

Figure 7.1. Comparison between peak low-back compression estimates made using a polynomial approach (POLY, McGill et al. 1996b) and an EMG-assisted musculoskeletal model (EMG, Cholewicki and McGill 1996). Data used to produce these estimates were collected from the 10 male subjects who performed the laboratory-simulated firefighting tasks described in Chapter 3.

Although justification was provided above for using the polynomial to calculate low-back compression forces during task performance, no attempt was made to incorporate the effects of trunk muscle activation and lumbar posture on low-back shear force magnitudes in Chapters 4, 5, or 6. Influenced by the findings of Norman et al. (1998) and since low-back shear force estimates are more sensitive to modeling assumptions than are low-back compression force estimates (Dieën and Kingma 2005), it was decided that reaction shear forces would constitute meaningful, albeit limited, information with respect to the thesis objectives. Norman et al. (1998) found that peak low-back reaction shear force estimates in the workplace are often better able (than are peak compression force estimates) to distinguish between individuals with and without a history of low-back pain. The authors convincingly argued that, in most cases, reaction shear loading patterns are directly related to the level of mechanical demands imposed on the low-back because it is unusual to observe increases/decreases in reaction force magnitudes without detecting corresponding increases/decreases in joint compression. However, there is one obvious exception (pure axial loading) where low-back mechanical demands would not necessarily correspond to reaction shear force magnitudes. To account for this special case in Chapters 4, 5, and 6, the angle of the lumbar spine was recorded when peak low-back compressive forces were imposed. Not only did documenting this variable more completely capture the low-back loading response to fireground task demands, but it could also be used to propose hypotheses about firefighter low-back injury potential given links between lumbar spine orientation and its capacity to support and transmit applied loads (Gallagher et al. 2005; Gunning et al. 2001; Howarth and Callaghan 2011).

Low-Back Loading and Hypothesized Injury Potential

Since it is not yet possible to validate musculoskeletal model outputs directly (i.e., measures of *in vivo* tissue forces cannot normally be acquired) and because outputs can be extremely sensitive to inherent model assumptions (Dieën and Looze 1999), experiments in Chapters 4, 5, and 6 were designed to make relative comparisons in the peak low-back loading responses to job demands. These studies aimed to test the effects that personal characteristics could have on the “direction” of peak low-back loading responses rather than on the “absolute” magnitudes of load estimates. For this reason, low-back load estimates were seldom discussed with reference to recommended exposure limits in Chapters 4, 5, and 6, except when used to decide if statistically significant differences were of biomechanically meaningful magnitudes. Nevertheless, based on discussion points raised throughout this thesis, it is argued that the peak low-back compression loading responses to simulated fireground task demands were reasonably well-approximated in Chapter 6. If so, results support the popular contention that fireground exposures could indeed be hazardous for low-back health, as peak low-back compression forces exceeded the NIOSH action limit of 3.4 kN in 80% of the simulated fireground tasks. When using the sledgehammer to simulate forcible entry and overhead chopping tasks in Chapter 6, peak low-back compression forces exceeded the NIOSH maximum permissible limit of 6.4 kN. Given that low-back overexertion injuries are commonly reported during fireground activities (Walton et al. 2003), it was not surprising that recommended exposure limits were exceeded during task simulations. However, no previous attempts had been made to quantify peak low-back loading levels during the performance of these tasks. As stated previously, the primary benefit gained from knowledge of the peak low-back load magnitudes is that risk assessments can be performed and thus priorities for intervention can be identified. For instance, based on the results in Chapter 6, firefighters who are older, de-conditioned, or have a low-back injury history could be discouraged from using sledgehammers (as tested) and trained to use alternative tools or techniques, where possible, to reduce the potential for (re-)injuring the low-back.

Though it is important to emphasize that data were not acquired from firefighters in Chapter 3 (subjects were student volunteers who passed the CPAT), additional insights were gained when a more sophisticated musculoskeletal modeling approach was employed. For instance, it was found in Chapter 3 that the peak low-back loading response varied considerably with inter-individual differences in body sizes and movement strategies. One way to interpret this finding is that there may be opportunities to devise movement-based training interventions to reduce the cumulative low-back load exposures by attenuating peak low-back loading during at least some of the fireground activities performed. There were several simulated tasks (e.g., victim rescue, forcible entry, equipment lift) wherein the external task demands predominately dictated the peak low-back loading response (i.e., recommended exposure limits were exceeded in nearly all subjects). But, there were other tasks (e.g., kneeling hose pull, hose advance) during which 50% or more of the subjects avoided potentially hazardous peak low-back loading levels. If individuals could be successfully trained to reduce cumulative low-back loading on the fireground, the ability to withstand peak low-back loading levels would conceivably be enhanced. The challenge, of course, is to impart spine "load-sparing" movement strategies that do not limit performance outcomes (e.g., sledgehammer impact force), as cumulative low-back loading would not necessarily be reduced (i.e., more time taken to meet task objectives) and the safety of the firefighters and victims could ultimately be compromised. Moreover, it remains to be seen if incumbent firefighters, with many years of experience and practice, could be trained to move and load their low-backs differently during fireground operations. Results of the study in Chapter 6 suggest that 12 weeks of general movement-centric exercise alone may be insufficient to alter habitual movement behaviour on-the-job. Perhaps more specific skill- or technique-focused instruction, provided when hired and over a longer time period, would be necessary to improve retention and transfer of training.

7.3. Future Research

Based on annual survey samples garnered by the National Fire Protection Association (NFPA), it is estimated that more firefighter musculoskeletal injuries are reported during fireground operations than in any other duty they perform (Karter and Molis 2010). And, as shown in Chapter 1 (Figure 1.1), the estimated number of musculoskeletal injuries reported per fire has doubled since 1981. However, the NFPA surveys also indicate that fireground musculoskeletal injuries may represent only 36.5% of all such firefighter injuries (Figure 7.2). In fact, results of several other studies suggest that more firefighter musculoskeletal injuries are amassed during on-duty exercise and training activities than during fireground operations (Bylund and Björnstig 1999; Loës and Jansson 2001; Poplin et al. 2011). Military service personnel also sustain many musculoskeletal injuries during exercise- and training-related activities (Evans et al. 2005; Heir and Glomsaker 1996; Gruhn et al. 1999; Jones and Knapik 1999; Kaufman et al. 2000; Lauder et al. 2000), and there is evidence to suggest that the number of such injuries can be reduced by manipulating the mode, frequency, duration, and intensity of military exercise and training regimens (Knapik et al. 2009). Therefore, future efforts to study the exercise and training practices of firefighters are warranted, especially since injuries sustained during these activities could conceivably be avoided and because these injuries may reduce the capacity to withstand demands associated with non-modifiable fireground tasks.

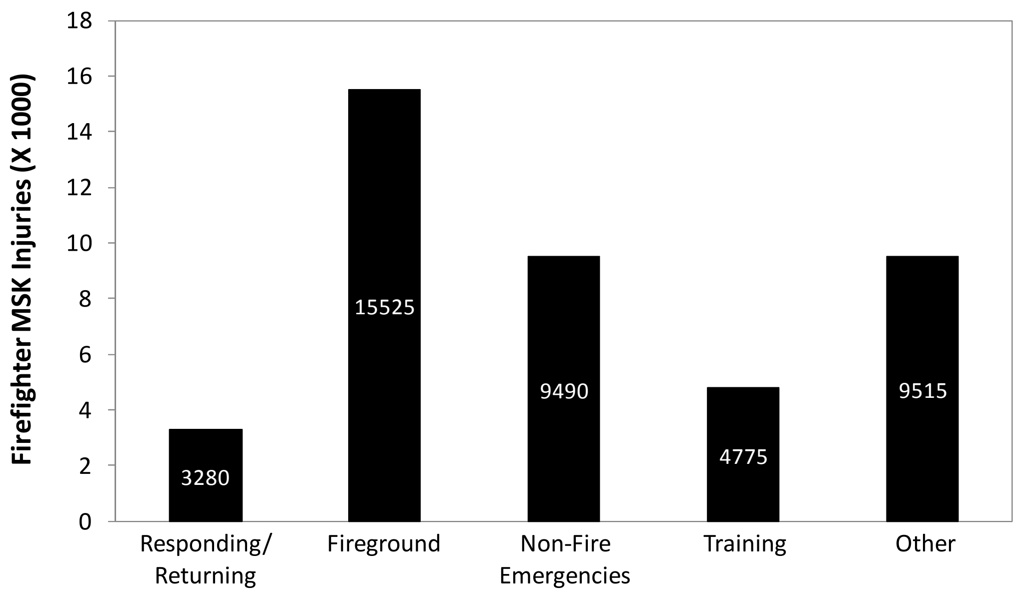

Figure 7.2. Total number of estimated line-of-duty musculoskeletal (MSK) injuries reported by American firefighters in 2009. Reported are estimates based on the results of the most recent survey data published by the National Fire Protection Association and projected to estimate national rates (Karter and Molis 2010).

It is also important to recognize that firefighters respond first to most medical emergencies, and thus many firefighters are cross-trained as paramedics. Consistent with the argument presented above, future success in preventing fireground overexertion injuries might be achieved by controlling low-back loading levels during non-fire emergencies. For example, in a collection of studies conducted by Conrad, Lavender, and colleagues (Conrad et al. 2000; Conrad et al. 2008; Lavender et al. 2007a; Lavender et al. 2007b; Lavender et al. 2007c), ergonomic hazards associated with firefighter patient handling duties were identified, biomechanical and postural risk analyses were performed, and results were used to conceptualize, develop and evaluate an ergonomic intervention to control peak low-back loading levels. If similar opportunities are exploited during other firefighter duties (e.g., equipment and tool handling and maintenance activities), it is possible that the number of fireground low-back overexertion injuries could be reduced. More research is needed to document cumulative low-back loading exposures in firefighters so that potential ergonomic hazards can be more fully appreciated and low-back injury prevention approaches can be more comprehensive.

7.4. Overall Conclusions

Results of this thesis confirm that activities performed on the fireground are potentially hazardous for low-back health, as peak low-back loading levels routinely exceeded recommended exposure limits during simulated task performance. This finding was anticipated given that low-back overexertion injuries are frequently reported during fireground operations. However, results of this thesis also indicated that inter-individual differences in movement strategies – related to personal characteristics such as body size, gender, and distal lower-extremity joint dysfunction – could alter occupational low-back loading demands and injury potential. It could not be concluded that occupational low-back loading demands and injury potential were consistently affected by short-term physical fitness improvements, nor could it be concluded that scoring above or below 14 on the Functional Movement Screen™ would project the peak low-back loading response to lifting. Future research should investigate the low-back mechanical demands associated with performing non-fireground duties, as opportunities may exist to implement ergonomic strategies to control cumulative low-back loading. Particular attention should be paid to the exercise and training practices of firefighters, as musculoskeletal injuries sustained during these activities are potentially avoidable and because they could reduce the capacity to withstand musculoskeletal demands imposed during non-modifiable fireground operations.

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APPENDICES

A.I. Trunk Muscle Activation Summary Tables (Chapter 3)

A.II. Functional Movement Screen™ – Description and Scoring

A.III. Exercise Program Templates

A.IV. Movement-Centric Exercise – Sample Instructions and Feedback Cues

A.V. Summary of Physical Fitness Test Results (Chapter 6)

APPENDIX I

A.I. Trunk Muscle Activation Summary Tables (Chapter 3)

Table A1. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Hose Advance (Initiation) task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	11.0 (4.3)	7.3 (1.9)	16.1 (2.3)	11.8 (1.9)	0.1796	0.0488	0.8799
		Left	21.0 (9.8)	11.4 (3.3)	25.3 (2.7)	19.3 (3.8)	0.4001	0.0571	0.6442
	Mean	Right	3.9 (1.8)	3.0 (1.0)	5.7 (0.8)	4.6 (0.6)	0.2796	0.0910	0.8845
		Left	6.6 (3.2)	3.7 (1.1)	10.4 (1.5)	7.9 (1.3)	0.1198	0.0484	0.8813
External Abdominal Obliques	Peak	Right	10.4 (2.5)	10.5 (2.9)	18.2 (2.7)	14.7 (1.6)	0.0646	0.3088	0.2906
		Left	19.6 (4.7)	20.3 (5.7)	31.8 (5.4)	26.6 (4.5)	0.1957	0.2975	0.1756
	Mean	Right	3.4 (0.8)	4.0 (1.1)	7.3 (1.3)	6.8 (0.8)	0.0166	0.8951	0.3832
		Left	6.0 (1.4)	6.9 (1.7)	11.8 (1.9)	10.8 (1.9)	0.0512	0.9417	0.2470
Internal Abdominal Obliques	Peak	Right	24.4 (9.7)	25.7 (7.9)	38.7 (5.1)	36.6 (6.5)	0.2137	0.9092	0.6781
		Left	46.5 (12.8)	35.0 (8.3)	64.5 (9.8)	58.6 (9.6)	0.1419	0.1096	0.5966
	Mean	Right	7.6 (2.3)	9.5 (2.5)	16.7 (1.9)	15.7 (2.4)	0.0193	0.7102	0.2604
		Left	14.4 (3.4)	14.2 (3.0)	25.1 (3.5)	22.0 (3.3)	0.0487	0.3736	0.4130
Latissimus Dorsi	Peak	Right	4.2 (1.2)	4.2 (1.1)	7.6 (2.5)	4.9 (0.9)	0.3163	0.2346	0.2312
		Left	23.2 (5.8)	20.9 (4.8)	19.5 (3.4)	25.6 (6.9)	0.9479	0.5127	0.1648
	Mean	Right	1.3 (0.4)	1.5 (0.2)	2.4 (0.5)	2.1 (0.4)	0.0813	0.7760	0.4863
		Left	7.3 (1.6)	9.0 (1.7)	7.7 (1.3)	10.1 (2.0)	0.7168	0.0469	0.7300
Thoracic Erector Spinae	Peak	Right	9.3 (2.7)	13.2 (4.0)	12.3 (2.8)	15.2 (3.8)	0.5321	0.2335	0.8570
		Left	19.8 (4.0)	18.4 (2.9)	24.7 (4.9)	29.0 (5.6)	0.2195	0.3675	0.0864
	Mean	Right	3.8 (1.2)	4.0 (0.7)	3.8 (0.9)	5.1 (0.9)	0.5982	0.3774	0.4982
		Left	7.6 (1.3)	8.6 (1.4)	9.0 (1.7)	10.2 (1.9)	0.4929	0.1718	0.9637
Lumbar Erector Spinae	Peak	Right	16.4 (5.1)	14.2 (3.3)	17.1 (2.9)	18.0 (3.6)	0.6681	0.6865	0.3146
		Left	16.6 (3.9)	19.1 (3.6)	27.0 (4.7)	31.8 (6.9)	0.0831	0.2449	0.7129
	Mean	Right	6.9 (2.0)	6.0 (1.1)	6.0 (1.1)	7.7 (1.4)	0.8346	0.6790	0.2072
		Left	5.3 (1.4)	7.4 (1.3)	7.7 (1.6)	9.9 (2.0)	0.2310	0.0574	0.9618

Table A2. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Hose Advance (Stride) task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	6.7 (2.6)	5.6 (1.9)	7.9 (2.5)	3.8 (0.8)	0.9153	0.0699	0.2823
		Left	8.4 (3.1)	6.7 (2.5)	12.3 (2.0)	5.9 (1.1)	0.6142	0.0039	0.0652
	Mean	Right	3.3 (1.1)	3.7 (1.3)	4.9 (1.6)	2.5 (0.4)	0.8969	0.1845	0.0746
		Left	4.2 (1.5)	4.3 (1.6)	7.4 (1.4)	3.6 (0.6)	—	—	0.0468
External Abdominal Obliques	Peak	Right	6.7 (1.5)	5.3 (1.2)	8.7 (1.8)	5.9 (0.9)	0.4460	0.0706	0.5257
		Left	12.3 (3.2)	14.2 (5.3)	17.2 (3.7)	9.5 (1.9)	0.9817	0.2490	0.0678
	Mean	Right	3.8 (1.1)	3.3 (0.8)	5.3 (1.1)	3.5 (0.5)	0.4384	0.1557	0.3958
		Left	7.9 (2.3)	9.4 (3.7)	11.7 (2.5)	6.4 (1.4)	0.9139	0.2895	0.0632
Internal Abdominal Obliques	Peak	Right	21.0 (8.3)	19.3 (5.5)	16.4 (2.3)	13.4 (2.6)	0.4646	0.3805	0.8080
		Left	37.7 (14.8)	31.6 (7.2)	37.8 (4.6)	25.6 (3.0)	0.7890	0.1023	0.5715
	Mean	Right	10.2 (3.7)	11.6 (3.1)	12.0 (1.8)	8.4 (1.8)	0.8496	0.4363	0.1010
		Left	24.9 (10.6)	19.1 (4.1)	26.0 (3.5)	17.0 (2.1)	0.9485	0.0980	0.7018
Latissimus Dorsi	Peak	Right	3.7 (1.1)	5.8 (3.0)	6.4 (2.2)	3.7 (0.8)	0.9100	0.8648	0.1576
		Left	19.3 (3.7)	18.6 (3.7)	21.4 (5.0)	22.5 (6.0)	0.6307	0.9342	0.7377
	Mean	Right	2.2 (0.6)	2.6 (1.0)	3.6 (1.3)	2.0 (0.4)	0.7351	0.4636	0.2027
		Left	12.3 (2.5)	12.2 (2.6)	14.9 (3.3)	15.7 (4.0)	0.4516	0.8506	0.8252
Thoracic Erector Spinae	Peak	Right	9.6 (2.3)	16.5 (6.1)	13.4 (2.4)	11.7 (2.1)	0.9086	0.4081	0.1810
		Left	24.2 (5.4)	21.2 (3.6)	22.8 (4.5)	22.1 (4.0)	0.9709	0.4763	0.6653
	Mean	Right	5.7 (1.7)	7.9 (2.7)	7.3 (1.4)	7.2 (1.5)	0.8324	0.4853	0.4724
		Left	15.5 (4.1)	14.2 (2.4)	14.9 (3.1)	15.0 (3.4)	0.9889	0.7552	0.7184
Lumbar Erector Spinae	Peak	Right	23.4 (5.5)	23.8 (5.3)	18.8 (2.7)	20.7 (3.7)	0.5215	0.5819	0.7238
		Left	21.9 (5.0)	24.1 (4.8)	21.6 (3.6)	23.1 (4.2)	0.9075	0.5440	0.8942
	Mean	Right	13.6 (4.2)	12.8 (3.9)	12.5 (1.9)	13.7 (3.1)	0.9825	0.9140	0.5798
		Left	13.1 (3.0)	14.1 (3.1)	13.8 (2.5)	15.5 (3.4)	0.7867	0.4865	0.8532

Table A3. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Kneeling Hose Pull task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	7.3 (2.2)	9.7 (2.3)	13.9 (5.4)	12.0 (2.4)	0.3015	0.9036	0.3465
		Left	7.4 (2.2)	10.7 (2.6)	14.9 (3.6)	17.6 (2.9)	0.0495	0.1903	0.8827
	Mean	Right	1.7 (0.6)	2.0 (0.6)	1.9 (0.5)	2.2 (0.3)	0.7463	0.2786	0.9206
		Left	0.9 (0.2)	1.9 (0.5)	2.5 (0.6)	2.8 (0.4)	0.0419	0.0330	0.2861
External Abdominal Obliques	Peak	Right	29.3 (6.7)	36.5 (11.2)	29.8 (6.8)	38.5 (5.9)	0.8945	0.1809	0.9020
		Left	19.5 (4.3)	27.3 (8.1)	30.6 (5.4)	32.2 (4.1)	0.2905	0.1639	0.3562
	Mean	Right	4.5 (1.3)	4.7 (1.4)	4.0 (0.8)	6.0 (0.7)	0.7900	0.2161	0.2975
		Left	3.0 (0.8)	4.6 (1.7)	4.3 (0.9)	5.0 (0.5)	0.5486	0.0614	0.4958
Internal Abdominal Obliques	Peak	Right	34.1 (13.2)	48.4 (18.3)	39.3 (9.5)	68.3 (13.4)	0.5160	0.0010	0.2002
		Left	45.9 (10.8)	59.2 (9.8)	67.7 (13.3)	70.3 (8.2)	0.2548	0.1827	0.3674
	Mean	Right	5.1 (2.3)	8.8 (3.8)	7.0 (1.8)	9.8 (1.9)	0.6794	0.0043	0.6449
		Left	7.2 (1.4)	9.4 (1.7)	13.0 (2.5)	12.1 (1.5)	0.1008	0.5334	0.1356
Latissimus Dorsi	Peak	Right	43.8 (7.8)	49.7 (8.9)	53.2 (9.4)	57.6 (8.2)	0.4669	0.1570	0.8426
		Left	58.5 (13.5)	65.0 (9.1)	65.2 (11.3)	73.9 (10.3)	0.6134	0.0825	0.7857
	Mean	Right	5.4 (0.9)	7.2 (1.4)	7.9 (1.0)	11.1 (1.7)	0.0690	0.0058	0.3642
		Left	8.1 (1.4)	9.7 (1.4)	10.3 (1.8)	12.2 (1.5)	0.2514	0.0612	0.8866
Thoracic Erector Spinae	Peak	Right	43.0 (9.0)	69.0 (20.2)	47.3 (4.6)	56.0 (5.4)	0.7435	0.1030	0.4013
		Left	40.0 (5.8)	43.6 (6.6)	40.9 (4.3)	49.9 (5.0)	0.6208	0.0515	0.3835
	Mean	Right	11.9 (1.5)	17.4 (4.6)	18.3 (1.6)	21.9 (2.1)	0.1069	0.0439	0.6598
		Left	11.2 (1.2)	12.6 (1.6)	15.1 (1.4)	16.5 (1.3)	0.0461	0.0669	0.9414
Lumbar Erector Spinae	Peak	Right	38.3 (4.8)	55.0 (8.0)	40.8 (3.3)	54.7 (3.8)	0.8786	<0.0001	0.6293
		Left	23.2 (2.6)	29.7 (3.2)	29.2 (2.0)	52.3 (6.6)	—	—	0.0359
	Mean	Right	17.6 (1.8)	19.9 (2.1)	20.2 (1.4)	25.8 (1.9)	—	—	0.0410
		Left	8.8 (1.6)	10.2 (1.6)	12.5 (1.0)	16.1 (1.3)	0.0148	0.0123	0.2134

Table A4. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Equipment Lift task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	7.3 (2.5)	8.2 (3.8)	6.9 (1.0)	7.8 (1.7)	0.8964	0.3608	0.9977
		Left	6.3 (1.1)	5.9 (1.3)	12.0 (2.6)	13.4 (4.7)	0.0819	0.7607	0.6318
	Mean	Right	2.5 (1.0)	3.0 (1.4)	2.0 (0.4)	2.2 (0.3)	0.5789	0.1319	0.5358
		Left	1.3 (0.2)	1.6 (0.3)	3.2 (0.7)	3.5 (0.9)	0.0318	0.2090	0.9042
External Abdominal Obliques	Peak	Right	8.5 (1.8)	8.0 (1.9)	7.6 (0.9)	10.7 (1.5)	0.6880	0.1587	0.0527
		Left	23.7 (5.8)	23.6 (6.1)	33.2 (4.3)	35.3 (4.4)	0.1586	0.5249	0.4914
	Mean	Right	2.0 (0.4)	2.1 (0.4)	2.2 (0.3)	3.2 (0.4)	0.2037	0.0145	0.0635
		Left	5.8 (1.4)	6.2 (1.5)	9.2 (0.9)	9.1 (1.3)	0.0953	0.7038	0.5158
Internal Abdominal Obliques	Peak	Right	24.8 (9.4)	26.0 (7.6)	35.8 (4.6)	38.0 (5.0)	0.2375	0.5339	0.8558
		Left	22.1 (9.6)	14.4 (3.3)	22.2 (2.5)	24.5 (2.8)	0.3460	0.6355	0.3766
	Mean	Right	7.6 (2.7)	7.0 (2.0)	10.4 (1.4)	11.1 (1.4)	0.2107	0.9620	0.4475
		Left	6.4 (2.7)	3.6 (0.7)	8.2 (0.7)	7.4 (0.8)	0.0747	0.2411	0.4850
Latissimus Dorsi	Peak	Right	28.1 (4.7)	28.7 (5.6)	31.1 (7.3)	42.9 (8.4)	—	—	0.0434
		Left	36.9 (6.2)	44.1 (7.7)	54.5 (7.2)	52.7 (7.0)	0.1859	0.3621	0.1366
	Mean	Right	6.6 (1.0)	8.1 (1.5)	9.2 (2.4)	12.0 (2.4)	0.2406	0.0006	0.2261
		Left	15.5 (2.8)	18.0 (3.3)	21.9 (2.8)	22.4 (2.7)	0.1889	0.1928	0.4058
Thoracic Erector Spinae	Peak	Right	43.5 (6.5)	48.3 (11.2)	51.9 (4.4)	64.3 (7.2)	0.2408	0.0558	0.3743
		Left	43.6 (5.8)	50.1 (7.8)	63.8 (3.6)	69.5 (5.1)	0.0139	0.1157	0.9091
	Mean	Right	14.6 (1.5)	18.3 (2.9)	18.8 (1.9)	23.6 (2.2)	0.1163	0.0023	0.6128
		Left	18.6 (2.2)	20.9 (2.7)	27.3 (1.7)	30.0 (2.3)	0.0093	0.0150	0.8423
Lumbar Erector Spinae	Peak	Right	52.6 (4.7)	57.3 (5.7)	52.2 (3.3)	77.5 (3.9)	—	—	0.0008
		Left	58.3 (7.3)	63.8 (6.2)	60.8 (4.7)	76.8 (6.3)	0.3445	0.0068	0.1548
	Mean	Right	18.3 (1.4)	24.6 (2.1)	20.9 (1.8)	29.6 (1.9)	0.1295	0.0000	0.2306
		Left	21.3 (2.6)	24.2 (2.2)	24.5 (1.7)	29.2 (2.3)	0.1848	0.0040	0.4335

Table A5. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Equipment Carry task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	3.9 (0.9)	4.2 (1.3)	2.4 (0.3)	3.0 (0.5)	0.2135	0.4469	0.7960
		Left	3.4 (0.9)	3.1 (0.7)	2.8 (0.5)	2.9 (0.4)	0.6304	0.8377	0.5477
	Mean	Right	2.2 (0.5)	2.8 (0.9)	1.8 (0.3)	1.9 (0.3)	0.3655	0.2925	0.4472
		Left	1.7 (0.4)	1.9 (0.4)	1.9 (0.5)	1.9 (0.3)	0.8314	0.6535	0.5749
External Abdominal Obliques	Peak	Right	5.0 (0.6)	4.7 (0.8)	5.2 (0.6)	7.7 (2.2)	0.3198	0.2187	0.1350
		Left	6.3 (1.2)	6.5 (2.1)	6.5 (1.0)	7.3 (1.3)	0.8053	0.4911	0.6343
	Mean	Right	2.6 (0.3)	2.7 (0.6)	3.5 (0.4)	4.5 (1.0)	0.0871	0.2340	0.3440
		Left	3.3 (0.7)	4.1 (1.6)	4.3 (0.7)	4.4 (0.8)	0.6195	0.4562	0.5849
Internal Abdominal Obliques	Peak	Right	17.4 (4.7)	15.4 (3.7)	14.2 (2.4)	14.9 (2.4)	0.6950	0.7016	0.4340
		Left	23.8 (5.8)	24.4 (4.6)	21.6 (3.4)	21.9 (4.9)	0.7066	0.8727	0.9547
	Mean	Right	7.9 (2.2)	9.2 (2.2)	9.4 (2.0)	8.0 (1.1)	0.9643	0.9663	0.1723
		Left	12.6 (2.9)	14.6 (2.9)	15.3 (2.9)	13.2 (2.7)	0.8598	0.9969	0.2470
Latissimus Dorsi	Peak	Right	9.8 (2.2)	14.2 (4.4)	10.8 (2.3)	9.1 (3.0)	0.5929	0.5575	0.1883
		Left	9.9 (1.3)	14.2 (2.7)	7.0 (1.5)	8.9 (2.4)	0.1258	0.0428	0.4203
	Mean	Right	6.9 (1.6)	9.4 (2.7)	7.0 (1.8)	5.4 (1.8)	0.4551	0.7223	0.1167
		Left	4.8 (0.7)	7.9 (1.9)	4.2 (0.9)	5.4 (1.2)	0.3170	0.0294	0.3249
Thoracic Erector Spinae	Peak	Right	13.3 (2.8)	18.9 (6.2)	13.9 (2.6)	15.3 (2.7)	0.7674	0.1431	0.3680
		Left	15.4 (3.2)	16.8 (3.0)	11.1 (1.7)	17.1 (2.8)	0.5626	0.0480	0.1941
	Mean	Right	8.2 (1.9)	11.2 (3.0)	9.3 (1.7)	9.2 (1.6)	0.8782	0.1304	0.1062
		Left	6.8 (1.4)	8.8 (2.1)	6.7 (1.3)	9.8 (1.6)	0.8290	0.0211	0.6225
Lumbar Erector Spinae	Peak	Right	16.1 (3.3)	22.1 (5.7)	12.7 (1.6)	17.6 (2.1)	0.4050	0.0176	0.8041
		Left	13.2 (1.7)	17.4 (2.8)	10.1 (1.0)	15.8 (2.1)	0.3739	0.0007	0.5645
	Mean	Right	7.6 (1.7)	12.1 (4.3)	7.8 (1.1)	9.7 (1.4)	0.7114	0.0834	0.4696
		Left	6.0 (0.9)	10.2 (2.4)	6.1 (0.9)	9.3 (1.4)	0.8242	0.0027	0.6180

Table A6. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Forcible Entry task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	43.6 (8.0)	28.3 (6.2)	49.2 (21.4)	24.9 (5.1)	0.9401	0.0506	0.6423
		Left	28.6 (6.4)	18.3 (4.5)	32.1 (5.9)	27.5 (5.1)	0.3585	0.0701	0.4741
	Mean	Right	7.3 (1.5)	6.2 (1.9)	6.0 (1.1)	4.5 (0.5)	0.4071	0.0455	0.7388
		Left	5.3 (1.0)	4.2 (0.9)	6.7 (0.9)	6.4 (1.1)	0.1830	0.0949	0.4176
External Abdominal Obliques	Peak	Right	52.1 (13.5)	52.9 (10.7)	50.3 (9.1)	44.7 (9.3)	0.7422	0.4692	0.3436
		Left	39.0 (7.4)	35.6 (8.6)	50.9 (6.4)	46.0 (4.6)	0.2526	0.0917	0.7501
	Mean	Right	11.6 (2.3)	11.8 (2.0)	9.9 (1.0)	9.0 (1.3)	0.3633	0.5643	0.4075
		Left	13.5 (3.6)	13.3 (4.0)	15.9 (2.3)	15.2 (1.9)	0.6227	0.4708	0.6956
Internal Abdominal Obliques	Peak	Right	66.9 (10.3)	66.3 (8.7)	84.7 (16.4)	84.2 (17.7)	0.3305	0.9457	0.9942
		Left	81.7 (21.1)	76.8 (15.6)	117.3 (27.6)	101.5 (24.5)	0.3533	0.0503	0.2782
	Mean	Right	16.6 (2.2)	16.8 (2.9)	26.1 (3.7)	24.0 (3.7)	0.0699	0.4642	0.3726
		Left	20.6 (3.6)	21.1 (3.2)	25.4 (3.2)	23.0 (3.3)	0.4796	0.2849	0.0889
Latissimus Dorsi	Peak	Right	36.5 (6.6)	36.3 (5.9)	44.3 (7.4)	44.3 (7.7)	0.4030	0.9704	0.9703
		Left	69.6 (7.7)	64.3 (6.6)	88.0 (8.0)	77.6 (10.5)	0.1686	0.0670	0.5386
	Mean	Right	8.1 (1.7)	8.8 (1.6)	12.9 (2.5)	12.6 (2.1)	0.1376	0.7784	0.4330
		Left	15.7 (1.6)	15.8 (1.9)	20.6 (2.0)	19.2 (2.4)	0.1427	0.4028	0.3738
Thoracic Erector Spinae	Peak	Right	49.0 (5.4)	69.2 (22.0)	50.3 (4.4)	57.0 (5.4)	0.6844	0.2101	0.5246
		Left	68.6 (8.8)	71.4 (12.0)	81.7 (9.4)	76.0 (8.7)	0.5060	0.7583	0.3668
	Mean	Right	14.1 (1.2)	22.0 (6.3)	18.9 (1.8)	22.6 (2.1)	0.5047	0.0679	0.4888
		Left	20.1 (2.7)	22.9 (2.9)	26.5 (2.8)	26.5 (2.6)	0.1980	0.1745	0.1831
Lumbar Erector Spinae	Peak	Right	72.3 (11.0)	72.2 (10.1)	49.6 (4.0)	59.8 (7.5)	0.1527	0.1117	0.1036
		Left	74.1 (9.5)	75.6 (7.8)	93.8 (11.8)	103.5 (12.1)	0.1182	0.0686	0.1724
	Mean	Right	18.1 (2.6)	21.9 (3.2)	17.1 (2.0)	21.3 (2.8)	0.8316	0.0003	0.8422
		Left	22.2 (3.3)	24.8 (3.2)	27.7 (2.7)	33.5 (3.7)	0.1229	0.0065	0.2642

Table A7. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Victim Rescue (Drag) task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	5.0 (1.1)	4.8 (1.1)	5.9 (0.8)	5.0 (0.5)	0.6756	0.1335	0.3534
		Left	3.4 (0.4)	3.3 (0.5)	6.3 (0.9)	5.5 (0.7)	0.0089	0.1857	0.2642
	Mean	Right	3.1 (0.8)	3.2 (0.9)	4.0 (0.6)	3.4 (0.3)	0.5476	0.3777	0.1399
		Left	1.9 (0.3)	2.2 (0.4)	4.4 (0.7)	4.0 (0.6)	0.0073	0.8493	0.1721
External Abdominal Obliques	Peak	Right	6.3 (1.5)	7.5 (1.6)	13.1 (2.1)	12.8 (3.1)	0.0525	0.6181	0.4439
		Left	6.2 (2.0)	7.0 (1.9)	11.3 (1.1)	10.3 (1.2)	0.0651	0.8110	0.1931
	Mean	Right	3.6 (0.9)	4.2 (0.9)	7.7 (1.4)	8.4 (1.9)	0.0341	0.2562	0.8873
		Left	3.1 (0.8)	3.5 (0.8)	6.5 (0.7)	6.4 (0.6)	0.0050	0.6544	0.3630
Internal Abdominal Obliques	Peak	Right	28.2 (10.0)	29.6 (6.8)	36.9 (4.5)	35.4 (7.0)	0.4527	0.9917	0.7258
		Left	28.0 (10.1)	22.3 (2.7)	47.5 (8.7)	36.6 (8.1)	0.1056	0.1247	0.6199
	Mean	Right	13.8 (4.7)	15.4 (3.6)	20.0 (2.6)	21.4 (4.4)	0.2539	0.4657	0.9769
		Left	14.2 (4.7)	12.6 (2.2)	21.8 (2.5)	18.6 (2.5)	0.0675	0.3978	0.7676
Latissimus Dorsi	Peak	Right	32.9 (5.2)	35.7 (5.9)	48.3 (7.7)	39.4 (6.2)	0.2720	0.2990	0.0541
		Left	33.0 (5.2)	42.1 (7.2)	47.0 (7.4)	34.8 (5.3)	—	—	0.0010
	Mean	Right	21.7 (3.3)	23.1 (3.6)	30.7 (4.5)	26.2 (4.6)	0.2872	0.3179	0.0678
		Left	21.0 (3.2)	27.1 (4.4)	29.0 (4.2)	23.0 (3.6)	0.0000	0.0000	0.0043
Thoracic Erector Spinae	Peak	Right	34.6 (4.1)	40.6 (6.9)	46.3 (5.4)	44.2 (6.3)	0.3446	0.4092	0.0889
		Left	28.6 (3.0)	26.9 (4.2)	38.5 (4.8)	36.3 (4.5)	0.1075	0.2500	0.8746
	Mean	Right	22.8 (2.7)	25.9 (4.8)	33.9 (4.3)	33.7 (5.3)	0.1334	0.3572	0.2835
		Left	18.4 (2.3)	17.7 (2.8)	28.1 (4.2)	25.2 (3.7)	0.0714	0.1849	0.3967
Lumbar Erector Spinae	Peak	Right	35.0 (3.8)	39.2 (4.4)	49.2 (4.6)	47.6 (2.8)	0.0303	0.6471	0.3187
		Left	29.8 (3.6)	30.9 (3.2)	39.2 (4.5)	36.7 (4.4)	0.1801	0.6076	0.1831
	Mean	Right	24.5 (3.1)	25.6 (3.7)	35.5 (3.4)	34.7 (2.7)	0.0302	0.9398	0.5986
		Left	19.5 (3.0)	18.9 (2.6)	30.3 (4.2)	28.1 (4.2)	0.0565	0.2244	0.4629

Table A8. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Victim Rescue (Initiation) task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	6.8 (2.5)	5.7 (1.7)	6.0 (1.2)	6.1 (1.0)	0.9086	0.7146	0.7000
		Left	5.4 (2.4)	3.6 (0.6)	6.3 (1.4)	8.5 (2.1)	0.1445	0.9029	0.2281
	Mean	Right	1.6 (0.5)	2.0 (0.6)	1.5 (0.2)	1.8 (0.2)	0.7912	0.0452	0.8091
		Left	0.9 (0.2)	1.2 (0.2)	1.8 (0.4)	2.5 (0.6)	0.0396	0.0414	0.3095
External Abdominal Obliques	Peak	Right	6.7 (1.6)	3.9 (1.0)	8.6 (1.5)	9.8 (1.8)	0.0536	0.4699	0.0675
		Left	5.7 (1.7)	4.7 (2.0)	8.2 (1.1)	8.7 (1.1)	0.1220	0.6864	0.2991
	Mean	Right	1.4 (0.3)	1.3 (0.4)	2.0 (0.2)	2.6 (0.4)	0.0413	0.1445	0.0869
		Left	1.2 (0.3)	1.7 (0.9)	2.1 (0.4)	2.4 (0.3)	0.2346	0.3167	0.7236
Internal Abdominal Obliques	Peak	Right	20.6 (6.1)	16.7 (3.6)	32.1 (6.4)	31.2 (6.0)	0.1100	0.2803	0.4938
		Left	19.7 (5.9)	12.9 (2.2)	28.2 (3.7)	30.0 (6.0)	0.0266	0.5517	0.3084
	Mean	Right	4.8 (1.2)	5.6 (1.9)	8.0 (1.0)	7.5 (1.1)	0.1586	0.8497	0.3569
		Left	4.9 (1.2)	4.5 (0.9)	8.4 (0.9)	8.1 (1.3)	0.0173	0.6246	0.9455
Latissimus Dorsi	Peak	Right	34.1 (6.6)	40.9 (7.7)	66.0 (7.1)	79.1 (9.4)	0.0019	0.0760	0.5534
		Left	40.1 (6.3)	51.9 (5.6)	68.9 (5.5)	73.7 (8.5)	0.0043	0.1287	0.5090
	Mean	Right	7.1 (1.3)	9.6 (2.0)	13.7 (1.3)	16.1 (1.4)	0.0052	0.0023	0.9187
		Left	8.1 (1.1)	11.6 (1.3)	14.4 (1.2)	16.1 (1.7)	0.0059	0.0061	0.2940
Thoracic Erector Spinae	Peak	Right	64.4 (12.5)	61.2 (11.5)	66.7 (6.9)	78.7 (6.3)	0.4185	0.5131	0.2664
		Left	54.6 (8.8)	46.8 (7.5)	65.7 (4.3)	74.3 (6.5)	—	—	0.0049
	Mean	Right	21.1 (1.9)	24.1 (4.3)	27.3 (2.8)	28.8 (2.7)	0.1753	0.2751	0.7303
		Left	19.4 (2.3)	18.6 (2.9)	27.9 (2.3)	28.9 (2.5)	0.0127	0.9203	0.3497
Lumbar Erector Spinae	Peak	Right	73.3 (10.3)	63.6 (3.3)	67.7 (5.9)	94.6 (10.4)	—	—	0.0039
		Left	65.0 (7.0)	64.7 (6.3)	63.1 (5.0)	76.3 (7.7)	0.5752	0.0966	0.0821
	Mean	Right	24.9 (2.0)	25.6 (1.5)	27.2 (1.6)	32.2 (2.3)	0.0803	0.0246	0.0827
		Left	22.3 (2.0)	25.4 (2.6)	26.3 (1.9)	27.6 (2.5)	0.3171	0.0675	0.4241

Table A9. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Ceiling Breach task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	8.2 (1.9)	8.8 (2.1)	9.2 (2.8)	9.8 (1.8)	0.7312	0.7082	0.9867
		Left	6.0 (1.3)	5.7 (1.5)	11.0 (2.7)	12.2 (2.9)	0.0393	0.8120	0.6811
	Mean	Right	2.5 (0.5)	2.9 (0.9)	2.1 (0.5)	2.4 (0.3)	0.5504	0.2828	0.7798
		Left	1.7 (0.4)	1.7 (0.4)	2.6 (0.5)	3.2 (0.7)	0.0982	0.3479	0.3571
External Abdominal Obliques	Peak	Right	12.6 (3.5)	11.1 (2.3)	13.9 (1.9)	12.8 (2.9)	0.6427	0.5606	0.9367
		Left	8.3 (1.1)	12.0 (5.0)	21.7 (4.0)	14.2 (2.7)	0.0758	0.5014	0.0626
	Mean	Right	3.2 (0.5)	3.7 (1.1)	3.9 (0.6)	4.3 (1.0)	0.5179	0.5103	0.9959
		Left	3.1 (0.6)	3.2 (1.0)	5.3 (1.2)	4.7 (1.0)	0.1643	0.7107	0.5412
Internal Abdominal Obliques	Peak	Right	41.2 (7.3)	46.6 (12.4)	49.3 (8.8)	51.9 (8.8)	0.5885	0.5000	0.8095
		Left	38.7 (5.1)	37.0 (8.7)	40.0 (5.7)	35.3 (4.6)	0.9793	0.4018	0.6889
	Mean	Right	14.0 (2.4)	17.6 (4.0)	15.3 (2.4)	16.0 (2.1)	0.9642	0.2259	0.4034
		Left	14.3 (2.0)	14.9 (3.3)	13.5 (1.5)	13.3 (1.4)	0.6708	0.8766	0.7446
Latissimus Dorsi	Peak	Right	19.1 (4.0)	13.1 (2.5)	22.2 (4.0)	20.2 (3.0)	0.2356	0.1212	0.4284
		Left	18.6 (4.3)	17.1 (3.1)	20.0 (2.7)	19.6 (2.6)	0.6471	0.5801	0.7556
	Mean	Right	7.5 (1.9)	5.1 (1.0)	7.9 (1.6)	8.2 (1.3)	0.3847	0.1932	0.1056
		Left	6.9 (1.7)	7.2 (1.6)	6.0 (0.8)	6.3 (0.9)	0.5932	0.5911	0.9655
Thoracic Erector Spinae	Peak	Right	40.0 (2.2)	46.0 (10.1)	60.7 (7.7)	66.2 (9.8)	0.0506	0.3566	0.9652
		Left	42.9 (4.9)	40.6 (5.6)	45.6 (3.2)	54.7 (6.3)	0.1828	0.3966	0.1609
	Mean	Right	16.5 (1.3)	18.3 (3.4)	25.8 (3.0)	26.4 (3.8)	0.0366	0.5327	0.7663
		Left	16.8 (2.2)	16.3 (2.7)	18.6 (1.5)	21.8 (2.4)	0.2181	0.3373	0.2093
Lumbar Erector Spinae	Peak	Right	40.7 (5.7)	40.9 (4.7)	53.0 (5.4)	65.8 (9.2)	0.0276	0.2018	0.2153
		Left	38.7 (3.9)	41.8 (4.2)	43.6 (3.9)	62.6 (9.4)	0.0715	0.0343	0.1160
	Mean	Right	16.2 (2.4)	16.3 (2.0)	21.1 (1.9)	25.6 (2.3)	0.0141	0.1764	0.2013
		Left	14.0 (2.0)	15.7 (2.2)	18.0 (1.8)	22.7 (2.7)	0.0593	0.0409	0.3152

Table A10. Peak and mean trunk muscle activation levels (% MVIC) during the performance of the Ceiling Pull task in Chapter 3. Data presented are the mean (SEM) values across all men (N = 10) and all women (N =10).

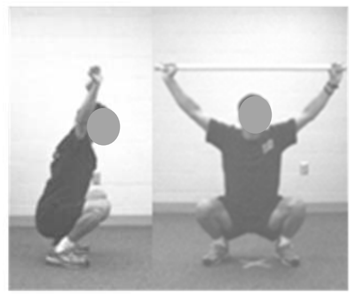
Muscle Group	Variable	Side	Males		Females		*p*-value		
			Pre	Post	Pre	Post	gender	time	gender×time
Rectus Abdominis	Peak	Right	132.0 (25.7)	108.7 (16.7)	107.5 (21.3)	82.4 (19.4)	0.3884	0.0056	0.9052
		Left	124.4 (25.3)	112.3 (28.6)	89.9 (11.7)	76.3 (10.6)	0.2235	0.1568	0.9290
	Mean	Right	32.3 (6.1)	27.8 (4.3)	24.6 (3.9)	20.7 (5.2)	0.5504	0.2828	0.7798
		Left	28.4 (5.6)	25.4 (4.7)	19.5 (2.4)	17.1 (2.7)	0.0982	0.3479	0.3571
External Abdominal Obliques	Peak	Right	69.9 (16.8)	59.7 (16.9)	49.9 (8.6)	46.8 (12.1)	0.4087	0.1300	0.4082
		Left	30.7 (8.7)	24.9 (4.3)	40.7 (6.2)	39.2 (9.4)	0.2234	0.4039	0.6275
	Mean	Right	17.8 (4.3)	18.1 (5.9)	16.6 (2.7)	15.4 (4.1)	0.5179	0.5103	0.9959
		Left	6.4 (1.3)	7.5 (1.4)	9.5 (1.5)	9.9 (2.5)	0.1643	0.7107	0.5412
Internal Abdominal Obliques	Peak	Right	83.6 (14.1)	74.4 (11.5)	78.2 (9.8)	74.4 (14.2)	0.8638	0.4385	0.7468
		Left	74.5 (17.7)	74.9 (16.3)	85.4 (12.4)	78.3 (18.2)	0.7495	0.6211	0.5822
	Mean	Right	23.9 (3.4)	24.5 (3.8)	27.4 (3.0)	26.6 (4.3)	0.9642	0.2259	0.4034
		Left	21.3 (3.8)	22.4 (4.1)	24.2 (2.4)	22.1 (3.9)	0.6708	0.8766	0.7446
Latissimus Dorsi	Peak	Right	57.8 (7.6)	40.7 (10.2)	74.1 (9.7)	48.7 (4.6)	0.2680	0.0007	0.4357
		Left	38.3 (4.8)	40.7 (7.4)	43.0 (10.4)	39.2 (5.6)	0.8657	0.8852	0.4966
	Mean	Right	15.8 (2.5)	12.8 (2.9)	20.7 (2.6)	14.4 (1.8)	0.3847	0.1932	0.1056
		Left	11.0 (1.5)	12.0 (2.2)	11.1 (1.9)	10.7 (1.7)	0.5932	0.5911	0.9655
Thoracic Erector Spinae	Peak	Right	45.1 (8.4)	36.8 (9.2)	47.6 (9.1)	37.6 (5.1)	0.8795	0.0316	0.8285
		Left	31.8 (6.0)	33.0 (5.5)	30.3 (2.7)	28.2 (3.9)	0.6104	0.8729	0.5585
	Mean	Right	14.4 (2.6)	14.2 (4.2)	13.6 (2.2)	12.1 (2.1)	0.0366	0.5327	0.7663
		Left	8.6 (1.4)	9.2 (1.4)	8.0 (0.8)	7.7 (0.9)	0.2181	0.3373	0.2093
Lumbar Erector Spinae	Peak	Right	23.3 (3.6)	25.4 (2.2)	35.1 (7.5)	38.2 (3.6)	0.0428	0.4544	0.8921
		Left	19.4 (3.7)	27.9 (5.1)	25.4 (4.5)	28.0 (5.0)	0.5840	0.1341	0.4108
	Mean	Right	6.2 (1.1)	7.4 (0.8)	8.0 (1.6)	10.5 (1.6)	0.0141	0.1764	0.2013
		Left	4.3 (0.9)	7.2 (1.3)	5.9 (1.5)	6.9 (1.6)	0.0593	0.0409	0.3152

APPENDIX II

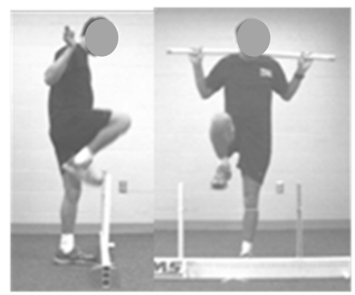
A.II. Functional Movement Screen™ – Description and Scoring

The Functional Movement Screen™ (FMS) was used as a tool in this thesis to identify personal movement qualities hypothesized to influence low-back loading and injury potential. The FMS is comprised of seven tasks which are purported to reveal overt limitations in active joint mobility, bilateral asymmetries, impaired static and dynamic postural control strategies, and pain-producing movement patterns (Cook et al. 2006a; Cook et al. 2006b; Cook et al. 2010). FMS tasks include: deep squat (DSQT); hurdle step (HSTP); in-line lunge (ILNG); shoulder mobility (SHLD); active straight-leg raise (ASLR); trunk stability push-up (PSHP); and rotary stability (RTRY). A four-point scoring system is used to rank FMS task performance. Scores range from 0 to 3, with 3 being the best score. If pain is perceived during task execution, a score of 0 is assigned. If an individual is able to execute an FMS task but must compensate (kinematically), a score of 2 is assigned. A score of 1 indicates that the FMS task cannot be executed as instructed, and a score of 3 indicates that the FMS task is performed exactly as instructed (i.e., without any observed movement compensations). Where applicable (HSTP, ILNG, SHLD, ASLR, RTRY), FMS tasks are performed bilaterally. If there is a bilateral asymmetry (i.e., right- and left-side scores are not equal), the lower score of the two sides is recorded. Three tasks (SHLD, PSHP, RTRY) include additional "clearing" movements, graded as positive or negative. Clearing movements are included to detect pain-producing movements or postures (e.g., spinal flexion/extension); if a positive clearing movement is noted ("pain"), then a score of 0 is recorded for the associated FMS task. If the clearance movement is negative ("no pain"), then the original FMS task score is retained. The total (composite) FMS score is calculated by adding the scores of individual FMS tasks; the best total score that can be attained on the FMS is 21 (perfect score = score of 3 × 7 FMS tasks). Specific FMS task instructions, ranking criteria, and examples are included below:

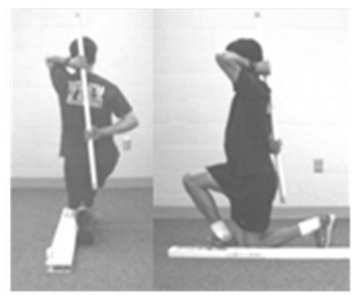
DEEP SQUAT (DSQT)

Test Instructions and Notes	Example	Grading Criteria	Example Scoring
Stand tall with feet shoulder-width apart and toes pointing forward		Femur below horizontal	√
Hold dowel with both hands and rest it on the head with elbows and shoulders at 90 degrees		Tibia/torso parallel	√
Press dowel straight overhead by straightening the arms		Knees aligned over toes	√
Squat down as deep as possible while keeping your torso upright, heels on the ground, and dowel in overhead position		Symmetrical weight-bearing	√
Pause at the deepest point and return to starting position		Dowel behind toes	√
Subject performs up to 3 times		Feet do not externally rotate	×
If subject does not score 3 (i.e., all grading criteria are not met), place a board under the heels and repeat task		Heels remain on floor	√
		Performs pain-free	√

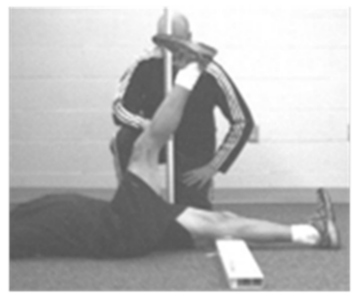
HURDLE STEP (HSTP)

Test Instructions and Notes	Example	Grading Criteria	Example Scoring
Stand tall with feet together and toes touching the board		Clears cord	√
Hold dowel with both hands and rest it behind the neck and across the shoulders		Hip, knee, ankle aligned	√
Remain standing tall, raise the right leg, step over the hurdle while keeping your foot, ankle, knee, and hip in alignment		Static spine posture maintained	√
Touch the heel to the floor, pause, and return to the starting position while keeping your foot, ankle, knee, and hip in alignment		Dowel remains parallel with ground	√
Score the right side		Ankle remains dorsiflexed	√
Repeat test with left leg		Maintains balance	√
Subject performs up to 3 times with each leg		Performs pain-free	√

IN-LINE LUNGE (ILNG)

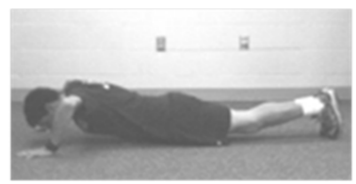
Test Instructions and Notes	Example	Grading Criteria	Example Scoring
Align dowel down the center of the back, ensuring that it touhes the back of head, upper back, and middle of buttocks		Dowel maintains points of contact	√
Hold the dowel in position with the left hand against back of neck and right hand against lower back		Dowel and torso remain vertical	√
Align left foot beside board with toe at the zero mark, align right foot beside the board with heel at the (*) mark		Static spine posture maintained	√
Keep both toes facing forward and feet flat		Knee touches ground behind heel	√
While standing tall and maintaining dowel contact with the head, upper back, and buttocks, descend until the left knee touches the ground directly behind the right heel, pause, and return to the starting position		Rear foot does not externally rotate	√
Score the right side		Maintains balance	√
Repeat test with legs and arms reversed		Front heel remains on ground	√
Subject performs up to 3 times with each leg		Performs pain-free	√

ACTIVE STRAIGHT-LEG RAISE (ASLR)



Test Instructions and Notes	Grading Criteria	Example Scoring
Lay flat on the back with backs of knees touching the board and toes pointing straight-up	Ankle resides between mid-thigh and ASIS	×
Align both arms next to the body with palms facing up	Ankle resides between mid-thigh and knee	√
Pull toes toward shins and keep both legs straight throughout	Ankle resides inferior to knee	×
Keep the back of the left knee in-contact with the board, raise the straightened right leg as high as possible, and pause for measurement	Contralateral hip remains neutral	√
Score the right side	Ankles remain dorsiflexed	√
Repeat test with left leg	Knees remain extended	√
Subject performs up to 2 times with each side	Performs pain-free	√

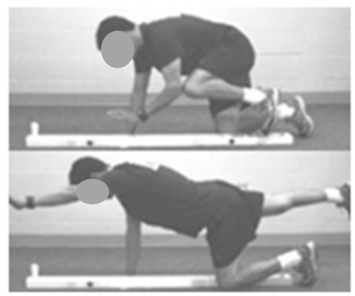
PUSH-UP (PSHP)

Test Instructions and Notes	Example	Grading Criteria	Example Scoring
Lie face-down with arms extended overhead and hands shoulder-width apart		Body moves in "one piece" with no spine motion	√
Move hands toward head until the thumbs are in-line with the forehead		Ankles remain dorsiflexed	√
Position legs together, pull toes toward shins, and lift elbows and knees off the ground		Performs pain-free	√
Keep the trunk rigid, toes pulled toward shins, and perform a push-up		No pain on spinal extension test	√
Subject performs up to 3 times			
If subject does not score 3 (i.e., all grading criteria are not met), slide hands until thumbs are in-line with chin and repeat test			

SHOULDER MOBILITY (SHLD)

Test Instructions and Notes	Example	Grading Criteria	Example Scoring
Stand tall with feet together and arms relaxed		Fists within one hand-length	×
With each hand, make a fist by wrapping fingers over thumbs		Fists within 1.5 hand-length	√
In one smooth and controlled motion, try and touch the fists together by reaching the right hand over the head and sliding it down the back while at the same time reaching the left hand behind and sliding up the back		Fists not within 1.5 hand-length	×
Do not wiggle or creep when attempting to touch fists; perform in one smooth motion and pause when the fists are as close together as possible		Performs pain-free	√
Measure distance between the fists (score the right side)		No pain on impingement test	√
Repeat test with arms reversed			
Subject performs up to 2 times with each side			

ROTARY STABILITY (RTRY)

Test Instructions and Notes	Example	Grading Criteria	Example Scoring
Get on "all-fours" with board between the hands and knees		Remains balanced	√
Keep hands under the shoulders and in-contact with the board and keep knees under the hips and in-contact with the board		Spine aligned with board	√
Pull toes toward shins and keep them in-contact with the board		Minimal transverse pelvis rotation	×
At the same time and in one smooth motion, reach the right hand forward and right leg backward (top picture), and without touching-down, touch the right knee and elbow together underneath the torso, pause, and return to extended position		Knee and elbow are touched	√
Subject performs up to 3 times on each side		Moving ankle is dorsiflexed	√
If subject does not score 3 on the ipsilateral version of the test (i.e., all grading criteria are not met), repeat the test in a contralateral fashion (i.e., right elbow touches left knee, and *vice versa*) (bottom picture)		Performs pain-free	√
		No pain on spine flexion test	√

APPENDIX III

A.III. Exercise Program Templates

FITNESS-CENTRIC EXERCISE TRAINING TEMPLATE

	PHASE 1				PHASE 2				PHASE 3			
DAY 1	**Week 1**	**Week 2**	**Week 3**	**Week 4**	**Week 5**	**Week 6**	**Week 7**	**Week 8**	**Week 9**	**Week 10**	**Week 11**	**Week 12**
1A. Trap Bar Deadlift	3 x 8	3 x 8	3 x 8	2 x 5	3 x 6	3 x 6	3 x 6	2 x 4	4 x 6	4 x 6	4 x 6	2 x 4
1B. Lat Pull-down/Pull-Up	3 x 8	3 x 8	3 x 8	2 x 5	3 x 6	3 x 6	3 x 6	2 x 4	4 x 6	4 x 6	4 x 6	2 x 4
1C. Bench Press	3 x 8	3 x 8	3 x 8	2 x 5	3 x 6	3 x 6	3 x 6	2 x 4	4 x 6	4 x 6	4 x 6	2 x 4
Rest:60s between sets												
2A. Dumbbell Military Press	2 x 10	2 x 10	2 x 10	1 x 6	3 x 10	3 x 10	3 x 10	2 x 6	3 x 8	3 x 8	3 x 8	2 x 5
2B. Dumbbell Bent Over Row	2 x 10	2 x 10	2 x 10	1 x 6	3 x 10	3 x 10	3 x 10	2 x 6	3 x 8	3 x 8	3 x 8	2 x 5
2C. Single Leg Squat	2 x 10	2 x 10	2 x 10	1 x 6	3 x 10	3 x 10	3 x 10	2 x 6	3 x 8	3 x 8	3 x 8	2 x 5
Rest:30s between sets												
3A. Leg Extension	2 x 15	2 x 15	2 x 15	1 x 15	2 x 10	2 x 10	2 x 10	1 x 10	2 x 8	2 x 8	2 x 8	1 x 8
3B. Hamstring Curl	2 x 15	2 x 15	2 x 15	1 x 15	2 x 10	2 x 10	2 x 10	1 x 10	2 x 8	2 x 8	2 x 8	1 x 8
3C. Abdominal Curl-Up	2 x 15	2 x 15	2 x 15	1 x 15	2 x 10	2 x 10	2 x 10	1 x 10	2 x 8	2 x 8	2 x 8	1 x 8
Rest:30s between sets												
CARDIO (run, bike, versa)	30 min LOW INTENSITY				30 min LOW INTENSITY				30 min LOW INTENSITY			
DAY 2	**Week 1**	**Week 2**	**Week 3**	**Week 4**	**Week 5**	**Week 6**	**Week 7**	**Week 8**	**Week 9**	**Week 10**	**Week 11**	**Week 12**
1A. Squat Press	2 x 15	2 x 15	2 x 20	1 x 12	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
1B. Horizontal Pull-Up	2 x 15	2 x 15	2 x 20	1 x 12	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
1C. Medicine Ball Slam	2 x 15	2 x 15	2 x 20	1 x 12	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
Rest: 45s between sets												
2A. Push-Up	2 x 15	2 x 15	2 x 20	1 x 12	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
2B. Lunge Walk	2 x 15	2 x 15	2 x 20	1 x 12	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
2C. Medicine Ball Rotation	2 x 15	2 x 15	2 x 20	1 x 12	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
Rest: 45s between sets												
3A. Grip (Squeeze)	2 x 15	2 x 15	2 x 20	1 x 20	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
3B. Wrist Roll	2 x 15	2 x 15	2 x 20	1 x 20	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
3C. Exercise Ball Crunch	2 x 15	2 x 15	2 x 20	1 x 20	2 x 25	2 x 25	2 x 30	1 x 20	2 x 35	2 x 35	2 x 40	1 x 25
Rest: 45s between sets												
CARDIO (run, bike, versa)	30 min MED INTENSITY (work:rest – 6:1 to 1:1)				30 min MED INTENSITY (work:rest – 6:1 to 1:1)				30 min MED INTENSITY (work:rest – 6:1 to 1:1)			
DAY 3	**Week 1**	**Week 2**	**Week 3**	**Week 4**	**Week 5**	**Week 6**	**Week 7**	**Week 8**	**Week 9**	**Week 10**	**Week 11**	**Week 12**
1A. Seated Leg Press	2 x 30s	2 x 30s	2 x 30s	1 x 30s	3 x 30s	3 x 30s	3 x 30s	2 x 30s	3 x 45s	3 x 45s	3 x 45s	2 x 30s
1B. Seated Chest Press	2 x 30s	2 x 30s	2 x 30s	1 x 30s	3 x 30s	3 x 30s	3 x 30s	2 x 30s	3 x 45s	3 x 45s	3 x 45s	2 x 30s
1C. Cable Row	2 x 30s	2 x 30s	2 x 30s	1 x 30s	3 x 30s	3 x 30s	3 x 30s	2 x 30s	3 x 45s	3 x 45s	3 x 45s	2 x 30s
Rest:45s between sets												
2A. Machine Squat	2 x 30s	2 x 30s	2 x 30s	1 x 30s	2 x 45s	2 x 45s	2 x 45s	1 x 45s	2 x 45s	2 x 45s	2 x 45s	1 x 45s
2B. Machine Shoulder Press	2 x 30s	2 x 30s	2 x 30s	1 x 30s	2 x 45s	2 x 45s	2 x 45s	1 x 45s	2 x 45s	2 x 45s	2 x 45s	1 x 45s
2C. V-Pulls	2 x 30s	2 x 30s	2 x 30s	1 x 30s	2 x 45s	2 x 45s	2 x 45s	1 x 45s	2 x 45s	2 x 45s	2 x 45s	1 x 45s
Rest:45s between sets												
3A. Biceps Curl	2 x 30s	2 x 30s	2 x 30s	1 x 30s	2 x 45s	2 x 45s	2 x 45s	1 x 45s	2 x 60s	2 x 60s	2 x 60s	1 x 60s
3B. Triceps Extension	2 x 30s	2 x 30s	2 x 30s	1 x 30s	2 x 45s	2 x 45s	2 x 45s	1 x 45s	2 x 60s	2 x 60s	2 x 60s	1 x 60s
3C. Side Plank	2 x 30s	2 x 30s	2 x 30s	1 x 30s	2 x 45s	2 x 45s	2 x 45s	1 x 45s	2 x 60s	2 x 60s	2 x 60s	1 x 60s
Rest:45s between sets												
CARDIO (run, bike, versa)	30 min HIGH INTENSITY (work:rest – 1:1 to 1:6)				30 min HIGH INTENSITY (work:rest – 1:1 to 1:6)				30 min HIGH INTENSITY (work:rest – 1:1 to 1:6)			

MOVEMENT-CENTRIC EXERCISE TRAINING TEMPLATE

	PHASE 1		PHASE 2				PHASE 3				PHASE 4	
DAY 1	**Week 1**	**Week 2**	**Week 3**	**Week 4**	**Week 5**	**Week 6**	**Week 7**	**Week 8**	**Week 9**	**Week 10**	**Week 11**	**Week 12**
1A. Upper Body Push	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	4 x 15	4 x 15	4 x 15	4 x 15	3 x 8	3 x 8
1B. Supplemental	N/A	N/A	3 x 5	3 x 5	3 x 5	3 x 5	3 x 5	3 x 5	3 x 5	3 x 5	3 x 5	3 x 5
1C. Lower Body Pull	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	4 x 12	4 x 12	4 x 12	4 x 12	3 x 8	3 x 8
1D. Supplemental Rest: 45s between sets	N/A	N/A	N/A	N/A	N/A	N/A	3 x 8	3 x 8	3 x 8	3 x 8	3 x 5	3 x 5
2A. Rotation	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	N/A	N/A	N/A	N/A	3 x 6	3 x 6
2B. Supplemental Rest: 45s between sets	2 x 6	2 x 6	2 x 5	2 x 5	2 x 5	2 x 5	N/A	N/A	N/A	N/A	2 x 5	2 x 5
3A. Upper Body Push	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	2 x 9	2 x 9
3B. Lower Body Pull Rest: 45s between sets	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	3 x 12	2 x 9	2 x 9
CARDIO (run, bike, elliptical)	30 min MED INTENSITY (low, mod and high HR)		30 min MED INTENSITY (low and high HR)				30 min MED INTENSITY (low, mod and high HR)				30 min MED INTENSITY (low, mod and high HR)	
DAY 2	**Week 1**	**Week 2**	**Week 3**	**Week 4**	**Week 5**	**Week 6**	**Week 7**	**Week 8**	**Week 9**	**Week 10**	**Week 11**	**Week 12**
1A. Lower Body Push	3 x 8	3 x 8	3 x 10	3 x 10	3 x 10	3 x 10	4 x 12	4 x 12	4 x 12	4 x 12	3 x 6	3 x 6
1B. Supplemental	N/A	N/A	3 x 5	3 x 5	3 x 5	3 x 5	3 x 6	3 x 6	3 x 6	3 x 6	3 x 5	3 x 5
1C. Upper Body Pull	3 x 8	3 x 8	3 x 10	3 x 10	3 x 10	3 x 10	4 x 12	4 x 12	4 x 12	4 x 12	3 x 6	3 x 6
1D. Supplemental Rest: 45s between sets	N/A	N/A	N/A	N/A	N/A	N/A	3 x 6	3 x 6	3 x 6	3 x 6	3 x 5	3 x 5
2A. Rotation	3 x 8	3 x 8	3 x 10	3 x 10	3 x 10	3 x 10	N/A	N/A	N/A	N/A	2 x 6	2 x 6
2B. Supplemental Rest: 45s between sets	2 x 6	2 x 6	2 x 8	2 x 8	2 x 8	2 x 8	N/A	N/A	N/A	N/A	2 x 6	2 x 6
3A. Lower Body Push	3 x 8	3 x 8	3 x 10	3 x 10	3 x 10	3 x 10	3 x 12	3 x 12	3 x 12	3 x 12	2 x 7	2 x 7
3B. Upper Body Pull Rest: 45s between sets	3 x 8	3 x 8	3 x 10	3 x 10	3 x 10	3 x 10	3 x 12	3 x 12	3 x 12	3 x 12	2 x 7	2 x 7
CARDIO (run, bike, elliptical)	30 min LOW INTENSITY (low and mod HR)		30 min LOW INTENSITY (low and mod HR)				30 min LOW INTENSITY (low and mod HR)				30 min LOW INTENSITY (low, mod and high HR)	
DAY 3	**Week 1**	**Week 2**	**Week 3**	**Week 4**	**Week 5**	**Week 6**	**Week 7**	**Week 8**	**Week 9**	**Week 10**	**Week 11**	**Week 12**
1A. Upper/Lower Body Push	3 x 10	3 x 10	3 x 10	3 x 10	3 x 10	3 x 10	4 x 10	4 x 10	4 x 10	4 x 10	3 x 6	3 x 6
1B. Supplemental	N/A	N/A	3 x 5	3 x 5	3 x 5	3 x 5	3 x 6	3 x 6	3 x 6	3 x 6	3 x 5	3 x 5
1C. Upper/Lower Body Pull	3 x 10	3 x 10	3 x 8	3 x 8	3 x 8	3 x 8	4 x 10	4 x 10	4 x 10	4 x 10	3 x 6	3 x 6
1D. Supplemental Rest: 45s between sets	N/A	N/A	3 x 5	3 x 5	3 x 5	3 x 5	3 x 6	3 x 6	3 x 6	3 x 6	3 x 5	3 x 5
2A. Rotation	3 x 10	3 x 10	3 x 8	3 x 8	3 x 8	3 x 8	N/A	N/A	N/A	N/A	3 x 6	3 x 6
2B. Supplemental Rest: 45s between sets	2 x 6	2 x 6	2 x 5	2 x 5	2 x 5	2 x 5	N/A	N/A	N/A	N/A	2 x 8	2 x 8
3A. Lower Body Push	3 x 8	3 x 8	3 x 8	3 x 8	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	2 x 7	2 x 7
3B. Upper Body Pull Rest: 45s between sets	3 x 8	3 x 8	3 x 8	3 x 8	3 x 8	3 x 8	3 x 12	3 x 12	3 x 12	3 x 12	2 x 7	2 x 7
CARDIO (run, bike, elliptical)	30 min HIGH INTENSITY (low, mod and high HR)		30 min HIGH INTENSITY (low and high HR)				30 min HIGH INTENSITY (low, mod and high HR)				30 min HIGH INTENSITY (low and high HR)	

APPENDIX IV

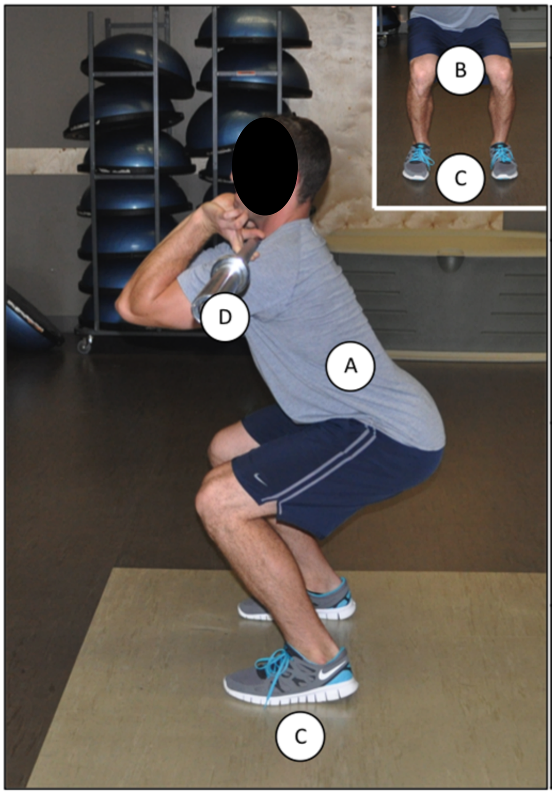
A.IV. Movement-Centric Exercise – Sample Instructions and Feedback Cues

In the pages that follow are examples of what was emphasized by the movement-centric exercise program coach (via instruction and feedback). The coach also relayed information in Table A.11 to help trainees understand why *Movement Matters!*

Table A.11. Common "weak links" associated with each fundamental movement pattern trained. Trainees were educated about how personal movement strategies increase/decrease musculoskeletal loading demands and increase/decrease musculoskeletal loading capacity.

Movement Pattern	Musculoskeletal Demand	Elevate Capacity	Reduce Capacity	Application to Firefighting
Squat	• Lower body - Knee (sagittal, frontal) • Trunk – Lower back (sagittal)	• Lumbar spine neutral (slight curve) • Knees aligned with feet and hips • Weight centered over mid-foot	• Hyperextend lumbar spine • Flex lumbar spine • Knees collapse medially or track laterally • Weight centered over toes (trunk upright)	• Operating/handling equipment • Search and rescue • Climbing
Lunge	• Lower body - Knee (sagittal, frontal, transverse) • Trunk – Lower back (sagittal)	• Lumbar spine neutral • Knees aligned with feet and hips • Weight centered over mid-foot	• Hyperextend lumbar spine • Flex lumbar spine • Knees collapse medially or track laterally • Weight centered over toes (trunk upright) • Feet not directed forwards	• Hose advance • Search and rescue • Climbing stairs/ ladder • Getting on/off truck
Lift	• Lower body - Knee (sagittal, frontal) • Trunk – Lower back (sagittal)	• Lumbar spine neutral • Knees aligned with feet and hips • Weight centered over mid-foot • Load kept close to body	• Hyperextend lumbar spine • Flex lumbar spine • Knees collapse medially or track laterally • Weight centered over toes (trunk upright) • Load not close to body	• Hoisting/carrying equipment • Victim rescue • Vehicle extrication
Push	• Trunk – Lower back (sagittal, frontal, transverse) • Upper body – Shoulders (sagittal, frontal, transverse)	• Head neutrally aligned • Scapular motion • Shoulders depressed • Lumbar spine neutral (no rotation)	• Head protrudes forwards • Shoulders rounded anteriorly • No scapular motion • Shoulders elevated • Hyperextend lumbar spine • Flex lumbar spine • Rotation in lumbar spine	• Ceiling breach • Hose advance • Forcible entry
Pull	• Trunk – Lower back (sagittal, frontal, transverse) • Upper body – Shoulders (sagittal, frontal, transverse)	• Head neutrally aligned • Scapular motion • Shoulders depressed • Lumbar spine neutral (no rotation)	• Head protrudes forwards • Shoulders rounded anteriorly • No scapular motion • Shoulders elevated • Hyperextend lumbar spine • Flex lumbar spine • Rotation in lumbar spine	• Hose drag • Pulling ceiling • Victim rescue

Squat Patterns – Coaching Points



Observation	Injury/Performance Considerations
A. Lumbar spine curvature	• Flexion or extension reduces the capacity of the spine. • Minimal spine motion (power) will increase the force applied to the external load.
B. Foot, knee and hip alignment	• Frontal plane knee motion reduces capacity. • To maximize effectiveness ground reaction forces should be directed through the knee joint.
C. Position of center of gravity (COG) relative to feet *(toe, mid-foot, heel)*	• Shifting the COG changes the amount of work done at each joint - towards the toe increases knee, towards the heel increases hip. • Seek to keep trunk parallel with shank.
D. Position of external load (if applicable)	• Distance between load and lumbar spine dictates work required at joint. • Minimize horizontal distance between the load (D) and COG (C) to maximize effectiveness

SQUAT PATTERNS

- Bodyweight squat
- Front squat
- Back squat
- Vertical jump

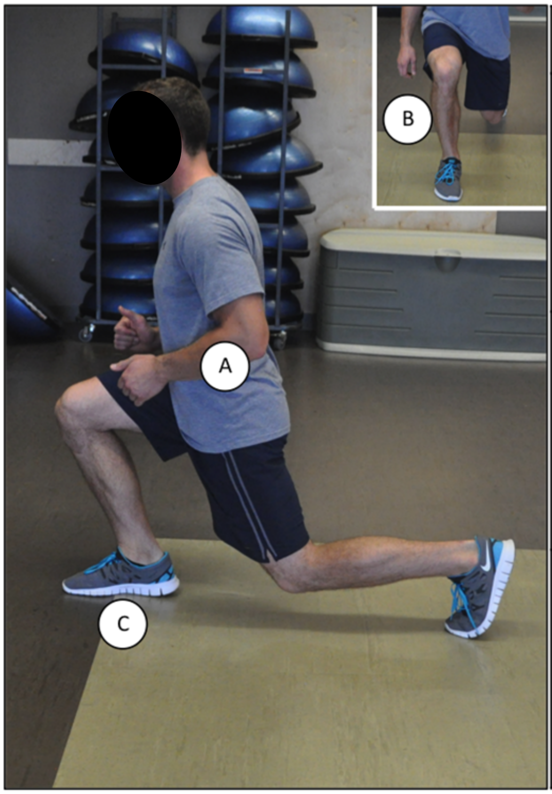
Common Compensations *(back and knees)*

- Lumbar extension
- Lumbar flexion
- Medial collapse of knees
- Weight on toes
- Weight on heels

Coaching Cues

- No spine motion (resist)
- Trunk and shins parallel
- Heels and toes on ground
- Bodyweight over mid-foot
- Grip ground with feet
- Barbell over feet
- Hips, knees, feet aligned
- Pull down, push up

Lunge Patterns – Coaching Points



Observation	Injury/Performance Considerations
A. Lumbar spine curvature	• Flexion or extension reduces the capacity of the spine, particularly under load. • Generating spine power (motion x load) is not an efficient means to improve performance.
B. Foot, knee and hip alignment	• Frontal plane knee motion reduces the joint's capacity. • Knee motion changes the direction of the ground reaction force and limits performance.
C. Position of bodyweight relative to front foot *(toe, mid-foot, heel)*	• The amount of work done at the ankle, knee and hip will change by varying the weight distribution on the front foot. • Towards the toe increases knee, towards the heel increases hip.

LUNGE PATTERNS

- Bodyweight lunge
- Split squat
- Back lunge
- Front lunge

Common Compensations *(back and knees)*

- Lumbar extension
- Lumbar flexion
- Medial collapse of knees
- Weight on front toe
- Hip/spine rotation

Coaching Cues

- No spine motion (resist)
- Trunk and shins parallel
- Front heel on ground
- Feet facing forwards
- Grip ground with front foot
- Hips, knees, feet aligned
- Pull down, push up

Lift Patterns – Coaching Points

Observation	Injury/Performance Considerations
A. Trunk angle versus spine posture	• Lumbar spine flexion under load will reduce capacity. • Provided that the trunk is stiffened and flexion is avoided a trunk lean can improve effectiveness.
B. Foot, knee and hip alignment	• Regardless of foot width, the feet, and knees should be aligned in the same direction. • Maximize effectiveness by directing force through joints in frontal plane.
C. Position of center of gravity (COG) relative to feet *(toe, mid-foot, heel)*	• Shifting the COG changes the amount of work done at each joint. Distribution is also dependent on the knee and trunk angle. • In general, target mid-foot.
D. Position of external load (if applicable)	• Minimize horizontal distance between load (D) and COG (C) to maximize effectiveness. • Distance between load and lumbar spine dictates work required of joint.

LIFT PATTERNS

- Deadlift
- Romanian deadlift (RDL)
- Single leg RDL

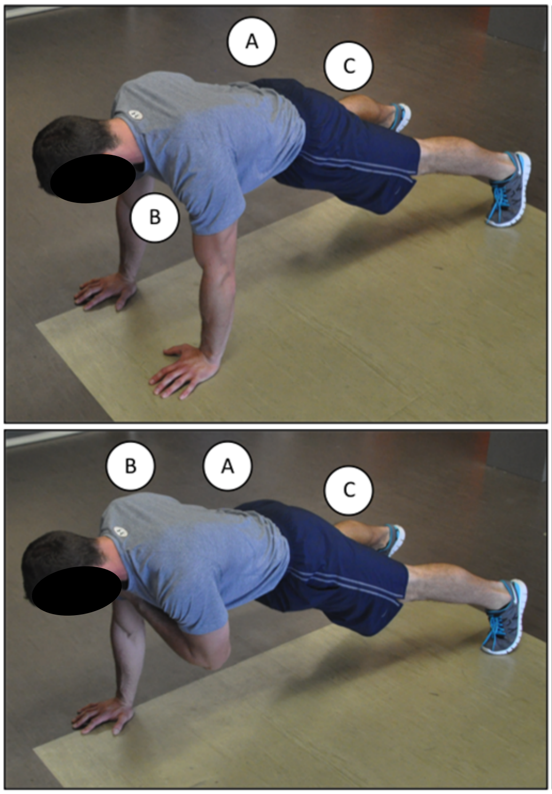
Common Compensations *(back and knees)*

- Lumbar extension
- Lumbar flexion
- Torso upright, on toes
- Shoulders behind load
- Load away from body

Coaching Cues

- No spine motion (resist)
- Heels and toes on ground
- Bodyweight over mid-foot
- Grip ground with feet
- Hips, knees, feet aligned
- Pull down, push up
- Shoulders aligned with load
- Keep load close

Push Patterns – Coaching Points



Observation	Injury/Performance Considerations
A. Lumbar spine curvature	• Flexion, extension or rotation can reduce the capacity of the lumbar spine and the efficiency of the movement • Spine motion may limit any benefit provided by the lower body.
B. Shoulder motion	• Anterior rotation and shoulder elevation can reduce the capacity of the joint. • Maximize effectiveness by controlling the motion of the shoulders and scapulae.
C. Use of lower body	• Every movement can be treated as a full body effort. • Maximize efficiency and effectiveness by integrating the lower body into every motion considered to be an upper body effort.

PUSH PATTERNS

- Push-up
- Bench press
- Standing press
- Unilateral press

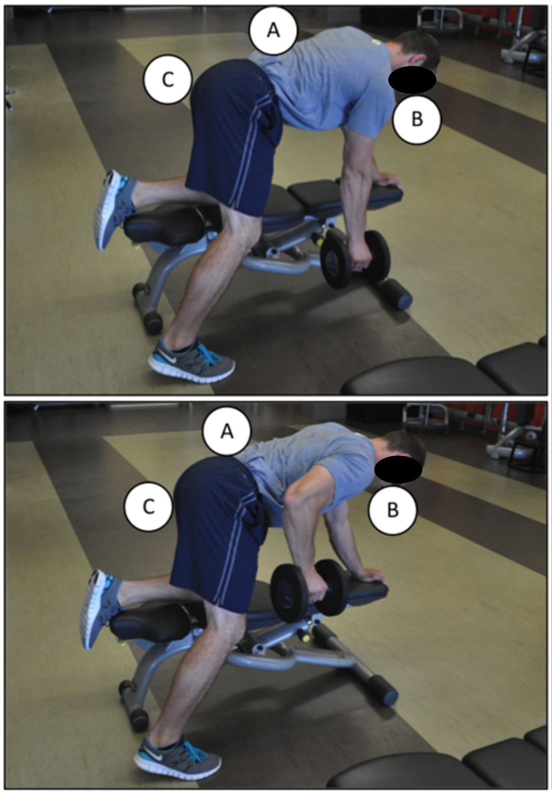
Common Compensations *(back and shoulders)*

- Lumbar extension
- Lumbar flexion
- Lumbar rotation
- Anterior rotation shoulder
- Shoulder elevation

Coaching Cues

- Head and chin back
- No spine motion (resist)
- Shoulders back and down
- Scapular motion
- Strong grip
- Pull load towards, push body away
- Use lower body

Pull Patterns – Coaching Points



Observation	Injury/Performance Considerations
A. Lumbar spine curvature	• Flexion, extension or rotation can reduce the capacity of the lumbar spine and the efficiency of the movement. • Spine motion may limit any benefit provided by the lower body.
B. Shoulder motion	• Anterior rotation and shoulder elevation can reduce the capacity of the joint. • Maximize effectiveness by controlling the motion of the shoulders and scapulae.
C. Use of lower body	• Every movement can be treated as a full body effort. • Maximize efficiency and effectiveness by integrating the lower body into every motion considered to be an upper body effort.

PULL PATTERNS

- Horizontal pull-up
- Pull-up
- Bilateral row
- Unilateral row

Common Compensations *(back and knees)*

- Lumbar extension
- Lumbar flexion
- Lumbar rotation
- Anterior rotation shoulder
- Shoulder elevation

Coaching Cues

- Head and chin back
- No spine motion (resist)
- Shoulders back and down
- Scapular motion
- Strong grip
- Externally rotate with pull
- Use lower body

APPENDIX V

A.V. Summary of Physical Fitness Test Results (Chapter 6)

Table A.12. Summary of physical fitness results in Chapter 6 for subjects in the control group (CON), fitness-centric exercise group (FIT), and movement-centric exercise group (MOV).

Physical Fitness Measure	**CON**				**FIT**				**MOV**				***p*-value**†		
	Pre		**Post**		**Pre**		**Post**		**Pre**		**Post**		**group**	**time**	**group×time**
Body Mass (kg)	93.0	(3.9)	92.7	(3.9)	95.1	(3.0)	94.7	(2.9)	94.8	(3.1)	95.0	(2.8)	0.8781	0.5231	0.7453
Body Fat (%)	18.7	(1.8)	18.9	(1.8)	18.5	(1.8)	17.1	(1.5)	16.8	(1.8)	15.4	(1.4)	—	—	0.0408
Treadmill Time (s)	663	(24.3)	640	(21.7)	665	(30.3)	749	(25.2)	640	(26.2)	703	(25.5)	—	—	< 0.0001
Predicted $V0_2$ max	39.4	(1.33)	38.4	(1.18)	38.8	(1.63)	42.9	(1.45)	38.5	(1.35)	41.4	(1.27)	—	—	< 0.0001
Trunk Flexion Endurance (s)	80.4	(13.6)	86.0	(11.5)	75.9	(10.8)	133.9	(11.5)	91.2	(12.7)	134.0	(11.4)	—	—	0.0007
Trunk Extension Endurance (s)	89.9	(9.7)	83.9	(9.0)	73.0	(4.8)	118.3	(8.9)	94.5	(10.9)	126.9	(10.1)	—	—	< 0.0001
Trunk Right Lateral Bend Endurance (s)	51.4	(6.6)	47.4	(3.8)	54.8	(7.3)	68.5	(3.7)	68.7	(9.5)	62.7	(4.3)	0.1083	0.8453	0.2279
Trunk Left Lateral Bend Endurance (s)	65.4	(9.2)	54.6	(5.2)	53.6	(5.6)	78.4	(5.0)	65.2	(7.4)	74.3	(3.9)	—	—	0.0114
Push-Ups (#)	39.1	(3.4)	43.6	(3.2)	39.4	(4.2)	65.5	(5.0)	36.8	(3.4)	50.3	(4.1)	—	—	< 0.0001
Keiser Chest Press @ 30 lb (W)	315	(9.7)	330	(11.8)	322	(12.6)	364	(17.4)	317	(11.2)	332	(14.7)	—	—	0.0392
Keiser Chest Press @ 50 lb (W)	375	(12.1)	381	(15.2)	408	(20.7)	416	(15.1)	372	(17.6)	386	(18.9)	0.2761	0.0675	0.7790
Keiser Chest Press @ 70 lb (W)	407	(16.5)	408	(16.4)	435	(30.4)	447	(20.4)	384	(22.2)	418	(24.5)	0.3968	0.0379	0.1535
Keiser Chest Press @ 90 lb (W)	408	(17.6)	414	(20.3)	423	(28.5)	456	(23.5)	393	(24.8)	434	(34.0)	0.6950	0.0015	0.1679
Keiser Chest Press @ 110 lb (W)	387	(20.2)	383	(22.3)	400	(30.9)	445	(25.8)	362	(28.7)	408	(38.3)	—	—	0.0305
Vertical Jump (cm)	54.3	(2.1)	54.8	(1.9)	53.9	(2.4)	57.0	(2.2)	54.1	(2.3)	56.7	(2.2)	—	—	0.0373
Keiser Squat @ 40 lb (W)	187	(10.2)	219	(12.0)	196	(12.9)	301	(13.1)	199	(11.2)	250	(15.0)	—	—	0.0018
Keiser Squat @ 60 lb (W)	347	(16.3)	376	(17.1)	374	(17.7)	470	(18.3)	359	(18.0)	425	(19.2)	—	—	0.0428
Keiser Squat @ 90 lb (W)	602	(24.0)	617	(18.6)	619	(21.6)	745	(24.7)	614	(29.1)	692	(29.3)	—	—	0.0032
Keiser Squat @ 120 lb (W)	832	(28.6)	861	(27.5)	865	(27.5)	1035	(29.0)	827	(33.9)	952	(32.9)	—	—	0.0002
Keiser Squat @ 150 lb (W)	1064	(34.6)	1082	(31.0)	1090	(30.8)	1269	(35.6)	1045	(44.8)	1170	(41.4)	—	—	0.0003
Right Grip Strength (kg)	46.6	(1.7)	48.3	(1.5)	48.0	(1.3)	49.7	(1.1)	46.6	(1.8)	48.8	(1.8)	0.7314	0.0012	0.7989
Left Grip Strength (kg)	45.8	(1.7)	47.6	(1.6)	45.4	(1.2)	47.4	(1.0)	44.7	(1.9)	46.7	(1.9)	0.9993	< 0.0001	0.9864
Sit-and-Reach (cm)	21.2	(2.4)	19.7	(2.0)	22.0	(1.9)	21.7	(1.9)	20.2	(2.1)	24.4	(1.5)	—	—	< 0.0001

†General linear model ANOVAs with one between-subject factor (group: CON vs. FIT vs. MOV) and one within-subject factor (time: Pre vs. Post) were performed to examine the impact of exercise on peak L4/L5 compression forces during task execution.