

Spinal Loads in Team Lifting

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Doctor of Philosophy**

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Abstract

The general purpose of this study was to examine factors that influence spinal loads during two-person team lifting in order to identify the spinal injury risks associated with using team lifting as a manual handling strategy. Three team lifting experiments were conducted, from which kinetic hand force data were obtained via boxes instrumented with force transducers and sagittal plane segmental kinematics acquired via a video based motion measurement system. Spinal loads (L4/L5 torque, compression and shear force) were subsequently calculated using a biomechanical model of the lumbar spine with a single equivalent extensor muscle force.

In the first experiment, spinal loads during individual and two-person team lifting tasks were compared. Ten healthy male subjects performed symmetrical individual lifts with a box mass of 15, 20 and 25 kg and symmetrical two-person team lifts with 30, 40 and 50 kg from the floor to standing knuckle height. Results indicated that the torque and compression force experienced by the lumbar spine were approximately 20% lower during team lifts compared to the load-matched individual lifting tasks. The two main and equal contributing factors reducing spinal load during team compared to individual lifting tasks were: (i) the increased horizontal pulling force and (ii) the ability of the team to hold the load at the ends of the box, which reduced the moment arm of the load.

The second experiment assessed the effect of relative team member height (matched versus unmatched) on lumbar spinal loads during two-person team lifting tasks. Twelve young healthy male subjects performed matched and unmatched team lifts with two box masses (30 and 60 kg) and three initial box heights (0, 20 and 40 cm). Matched team members had standing heights within 5%, whilst unmatched teams had an average standing height difference of 25 ± 2.5 cm. Although spinal loads were reduced for the shorter subjects and increased for the taller subjects at the end of the lift, no significant

difference was found in the maximum spinal loads incurred during matched compared to unmatched lifting conditions.

In the final experiment the relationship between load mass distribution and the relative spinal loads incurred by each of the individual team members during two-person team lifting tasks was examined. Two-person lifting teams were required to lift a box containing a mass of 30 kg or 60 kg from the floor to standing knuckle height. Adjusting the position of the centre of mass within the box by ± 15 cm and ± 7.5 cm relative to the evenly distributed position (0 cm) yielded three load mass distribution ratios (69:31, 59:41 and 50:50), which represented the percentage of the total mass lifted by each team member. Although the spinal load incurred by the team member lifting at the heavier end of the load was greater than for the person at the lighter end of the load, the difference between the spinal loads incurred by each team member was not as great as the difference in the asymmetric distribution of the load mass. Subsequent investigation of the factors influencing spinal load indicated that the spinal loads experienced by the team member at the heavier end of the load was less than expected because they generated a larger horizontal pulling force than their lifting partner. Consequently, during the lift the load translated toward the team member at the heavier end of the load, which combined with the larger horizontal pulling force reduced the extensor torque required at the lumbar spine.

Overall, results from this study have demonstrated that: (i) the lifting strategy used by two-person teams is distinguished from individual lifts by a greater use of horizontal pulling forces applied to the load and a decreased distance between the load and the lumbar spine, (ii) both the horizontal pulling force and the position of the hands on the load in team lifting have a load relieving effect on the lumbar spine and (iii) two-person team lifts performed by team members of unmatched standing height and with asymmetrical load mass appear to be coordinated in a manner that partially mitigates the increased spinal loads for the team member at increased risk of spinal injury.

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Statement of Originality

I certify that this thesis is my own work and to the best of my knowledge has not been submitted in part or in full to any academic institution for the award of any degree. Furthermore, all intellectual material contained within this document, except where due acknowledgments are made, is my own and has been produced as a consequence of the research conducted within this doctoral candidature.

.....

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Thesis Organisation

Chapter 1 provides a general introduction to the thesis and specifically states the research objectives.

Chapter 2 reviews the relevant literature, including: the incidence, cost and causes of LBP, the mechanisms of spinal injury, OHS guidelines for manual handling and research that has specifically investigated team lifting.

Chapter 3 details the techniques that have been used to measure spinal load and then describes and validates the biomechanical spinal model used in the three experimental studies presented in chapters 4, 5 and 6.

Chapter 4 describes an experiment that assessed spinal loads during individual and team lifting tasks. This chapter has been published in *Ergonomics*, 45(10), 671-681.

Chapter 5 describes an experiment that compared the spinal loads in two-person teams of matched versus unmatched standing height. This chapter has been published in the *International Journal for Industrial Ergonomics*, 32(1), 25-38.

Chapter 6 describes an experiment that examined the relationship between load mass distribution and the spinal load incurred by the members of a two-person lifting team. This chapter has accepted for publication in the *International Journal for Industrial Ergonomic*, (In Press).

Chapter 7 provides a summary and synthesis of the results from the three experiments presented in chapters 4, 5 and 6 and draws conclusions for the use of team lifting in the workplace based on the objectives stated in chapter 1. Recommendations for future research on team lifting and final conclusions are also made.

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List of Publications

The publications listed below are directly related to the research conducted as part of this thesis.

Published articles

Dennis, G.J. and Barrett, R.S. (2000). Ergowatch: a new manual handling management tool. *Ergonomics Australia*, 14(1), 17-27.

Dennis, G.J. and Barrett, R.S. (2002). Spinal loads during individual and team lifting. *Ergonomics*, 45(10), 671-681.

Dennis, G.J. and Barrett, R.S. (2003). Spinal loads during two-person team lifting: effect of matched versus unmatched standing height. *International Journal of Industrial Ergonomics*, 32(1), 25-38.

Dennis, G.J. and Barrett, R.S. (2003). Spinal loads during two-person team lifting: effect of load mass distribution. *International Journal of Industrial Ergonomics*, (In Press).

Conference proceedings

Dennis, G.J. and Barrett, R.S. (2000). Lumbar spinal loads in a tyre-lifting task: An application of Ergowatch. *Proceedings of the 3rd Australasian Biomechanics Conference*, Gold Coast: Australia, pp. 21-22.

Dennis, G.J. and Barrett, R.S. (2000). Recent developments in applied spinal modelling. *Proceedings of the 5th Biennial Motor Control & Human Skill Research Workshop*, Gold Coast: Australia, pp. 55.

Dennis, G.J. and Barrett, R.S. (2001). Lumbar spinal load in team versus individual lifting. *Proceedings of the 18th Congress of the International Society of Biomechanics*, Zurich: Switzerland, pp. 236.

Chapter 1

Introduction

1.1 Background

1.1.1 The lower back pain problem

Lower back pain (LBP) originating from occupational related spinal injuries is one of the foremost occupational health and safety (OHS) issues, with lifetime incidence rates of up to 85% (Spengler et al. 1986). In Australia, occupationally incurred LBP represents 25% of all workplace injuries (BLS, 2000), and results in worker compensation claims totalling \$AUD1.2 billion annually (NOHSC, 1996). Moreover, the financial cost of these claims does not include expenses incurred from lost productivity and staff retraining. LBP can also incur significant psychological and social costs, which although more difficult to calculate than fiscal costs, should not be underestimated (Lancourt and Kettelhut, 1992; Leamon, 1994). In the United States of America, the total economic burden imposed by the psychological, social and direct financial costs of occupationally related LBP has been estimated to be as high as \$US100 billion each year (NIOSH, 1996).

Due to the large economic costs associated with LBP, a considerable amount of research has been focused on determining the factors that contribute to LBP. Consequently, age, gender, strength, ethnicity, stress, socio-economic status, whole body vibration, peak and cumulative spinal loading, environmental conditions and physical fatigue have all been examined for possible links to LBP. However, of all the factors that have been investigated, manual handling and particularly lifting, has been the most commonly implicated cause of LBP (Jensen, 1988; NIOSH, 1997; Burdorf and Sorock, 1999). Specifically, the presence of excessive spinal load during lifting has been linked to

spinal injury that results in episodes of LBP (McGill, 1997; Lavender et al., 2003). Accordingly, many studies have examined various lifting tasks with the aim of identifying lifting techniques or practices that reduce the loads on the lumbar spine and thus decrease the risk of spinal injury and LBP.

1.1.2 Spinal load as a risk factor for spinal injury

The mechanical function of the spine is to allow movement of the trunk whilst providing enough support to maintain its structural integrity (Cholewicki and McGill, 1995). Thus, the spine is required to provide adequate sagittal, lateral and rotational movement whilst withstanding the compression and shear forces generated externally by inertial and weight forces and internally by the biological tissues in the lumbar region. If these forces on the lumbar spine exceed the structural capabilities of the biological tissues in the lumbar region spinal injury will occur (McGill, 1997; Neumann et al. 2001). Based on this hypothesis, it is generally accepted that excessive force (acute or cumulative) in the lumbar spine will result in damage to one or more of the spinal tissues that can lead to LBP and eventually occupational incapacitation. Furthermore, damage to one tissue can result in the overexertion of other spinal structures to compensate for the functional deficit, placing those tissues at increased risk of injury. Consequently, if work is continued further injury may occur, which can ultimately result in a cyclic process of chronic LBP (Figure 1.1).

Although major or permanent spinal injury can occur from a single excessive biomechanical loading event, LBP is more often caused by cumulative loads that occur as a result of the daily occupational stress placed on the lumbar spine (McGill, 1997; Norman et al., 1998). The mechanism of injury due to cumulative loading can be as a result of either cyclic loading of the spine during repetitive manual handling tasks, or the application of a sustained spinal load over a long duration. In either case, spinal

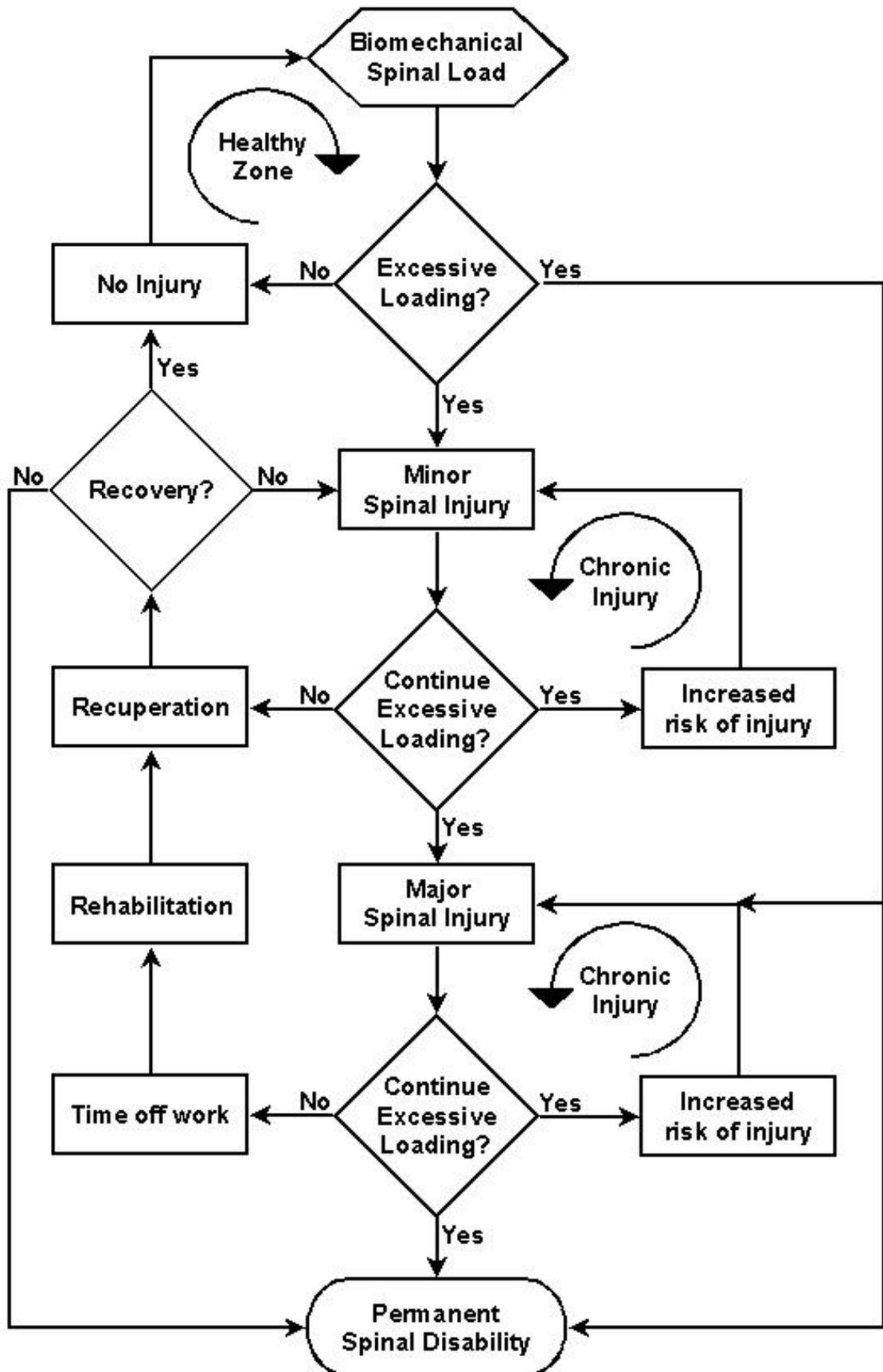


Figure 1.1: Model of the biomechanical origins of spinal injury.

injury occurs because the functional capacity of the spinal tissues is reduced as a result of prolonged spinal loading. Even forces well below the structural limits of the spinal tissues can reduce the structural capacity of those spinal tissues over time, and thus increase the risk of spinal injury in the future (McGill, 1997; Norman et al., 1998). Regardless of the cause of spinal injury (i.e. acute or cumulative spinal loading), if the force on the lumbar spine can be minimised then the risk of injury is also likely to be reduced (Figure 1.1). Therefore, in order to reduce the risks associated with manual handling and specifically lifting tasks performed in the workplace, most industrialised nations have implemented manual handling guidelines that are designed to reduce the risk of spinal injury by minimising the load experienced by the lumbar spine.

Australia, like most other industrialised nations, has a manual handling code of practice that follows a pattern of risk identification, assessment and control specific to each manual handling task. The guidelines for safe lifting in Australia, as set by the National Occupational Health and Safety Commission (NOHSC), are specified in the Manual Handling: National Code of Practice (NOHSC, 1990). Whilst this document also deals with the appropriate age, clothing and skill of employees and optimal visual and environmental conditions, the essence of the code is to reduce the acute and cumulative spinal loads experienced by employees performing manual handling tasks.

Due to the different size and shapes of objects lifted in the workplace, it is impossible to recommend a single lifting technique that is appropriate under all circumstances (Parnianpour et al., 1987; Burgess-Limerick et al., 1995; van Dieën et al., 1999). As a result, a generic set of lifting guidelines is recommended in order to reduce the loads on the lumbar spine. Typically, spinal load is minimised by lifting with a slow and controlled semi-squat technique whilst holding the load as close to the body as possible (NOHSC, 1990; Burgess-Limerick, 2003). When an individual cannot safely lift a load, the Manual Handling: National Code of Practice recommends that either an administrative intervention strategy (that removes the need for the lift to be performed)

or a mechanical lifting device be employed as the first method of risk control. In a situation where neither of these risk control strategies can be implemented then team lifting is advocated as an alternative method for reducing the load on the lumbar spine (NOHSC, 1990).

The guidelines on team lifting in the Australian Manual Handling: National Code of Practice state: *“To enable load sharing, lifting partners should be of similar height and build and should be trained in lifting techniques. There should be a person nominated as team leader to coordinate the lift. Team lifting should not be used as a first option in risk control”* (NOHSC, 1990). However, thus far the research that has been conducted on manual handling and specifically lifting has focused almost entirely on individual lifting tasks. Although some of the results from individual lifting research are applicable to team lifting, the dynamic interaction between team members may modify the spinal loads experienced by each team member and consequently affect the risk of spinal injury.

1.1.3 Team lifting research

Most team lifting studies to date have attempted to assess lifting performance by quantifying psychophysical variables. The psychophysical approach uses self-based estimates of maximum lifting capacity that are intended to reflect the ability of an individual or team to safely lift a load. The rationale underlying this form of lifting assessment is that the higher the self estimated maximum lifting capacity the greater will be the safety margin when lifting heavy loads. Most psychophysical team lifting studies (Kowalski, 1988; Kowalski and Pongpatanasuegsa, 1988; Rice et al., 1995) have found that the lifting capacity of a team is less than the summed lifting capacity of the individual members. However, contradictory results have also been reported (Johnson and Lewis, 1989; Mital and Motorwala, 1995). Psychophysical assessment criteria were also used by Lee and Lee (2001) to compare the lifting capacity of two-person teams

performing lifting tasks with team members of matched and unmatched standing height. Although the standing height difference between team members of unmatched height was only 4 cm, Lee and Lee (2001) reported significant increases in the maximum acceptable weight lifted by height matched compared to height unmatched teams.

An alternative to the psychophysical approach is the biomechanical approach, which unlike the psychophysical research paradigm specifically investigates the mechanical causes of spinal injury. The most recent and comprehensive biomechanical study that compared the spinal loads experienced in individual lifting tasks with those during team lifting tasks was conducted by Marras et al. (1999). Results from this study were not conclusive as lower spinal loads were found in symmetrical team lifting tasks and higher spinal loads in asymmetrical team lifting tasks, as compared to the spinal loads measured in corresponding load-matched individual lifting tasks. Mital and Motorwala (1995) measured spinal load as well as perceived lifting capacity in team lifting and found that when lifting a person access cover spinal loads were lower in a two-person team compared to individual lifts. However, this result is not really surprising considering the weight lifted was the same for both individual and team lifting tasks. Thus far no studies have sought to uncover the specific reasons why spinal loads are different during team compared to individual lifting tasks.

1.1.4 Horizontal hand force and spinal load in team lifting

There are many factors that affect the load experienced in the lumbar spine during lifting, including: load and body mass, horizontal distance the load is from the body and the vertical distance the load is moved. Research that has examined the effect of these factors on spinal load has subsequently led to formulation of the 1991 NIOSH lifting equation, which is extensively used as an ergonomic tool to estimate the risks associated with particular lifting tasks. However, one factor that is not considered by the 1991 NIOSH lifting equation is the horizontal force exerted by the hands upon the load.

The lumbar extension torque required during lifting is a function of the weight of the upper body and the load in the hands, and their respective moment arms. A decrease in the moment arm of the load will accordingly decrease the extension torque required at the lumbar spine. The moment arm of the load is the perpendicular distance between the line of action of the resultant force experienced by the hands (RHF) and the centre of rotation at the L4/L5 joint (Figure 1.2a). Pulling the load toward the body whilst lifting will generate a horizontal hand force (HHF) that causes the RHF to be directed away from the body (see Figure 1.2b). Whilst this change in RHF direction can also increase the magnitude of the RHF, it will decrease the moment arm of the load when the hands are vertically positioned below L4/L5 (Figure 1.2b). Thus, the flexion torque generated by the load will decrease, which reduces the compression force on the lumbar spine due to the lower extension torque required by the lumbar extensors. Furthermore, the moment arm of the RHF is most likely to be minimised at the start of the lift when the loads experienced by the lumbar spine are at a maximum.

During individual lifting, the contribution of the HHF to the RHF vector is relatively minor. A net HHF will result in an acceleration of the load that typically results in the translation of the object in the direction of the net HHF. Consequently, a large horizontal pulling force is difficult to maintain during individual lifting, especially if the load is horizontally located close to the lifter. Furthermore, a large HHF may compromise the balance of the lifter, especially if the coefficient of friction between the lifter and the floor is low. Consequently, the RHF is predominantly vertical during most individual lifting tasks and thus the moment arm of the load with respect to the lumbar spine can be quite substantial (Figure 1.2a).

In a team lifting task the direction of the RHF may be vastly different from the RHF during an individual lifting situation. Unlike individual lifting, the presence of a HHF will not translate the load during team lifting if the team members pull in equal but opposite directions. As a result, the RHF is directed away from the body and the

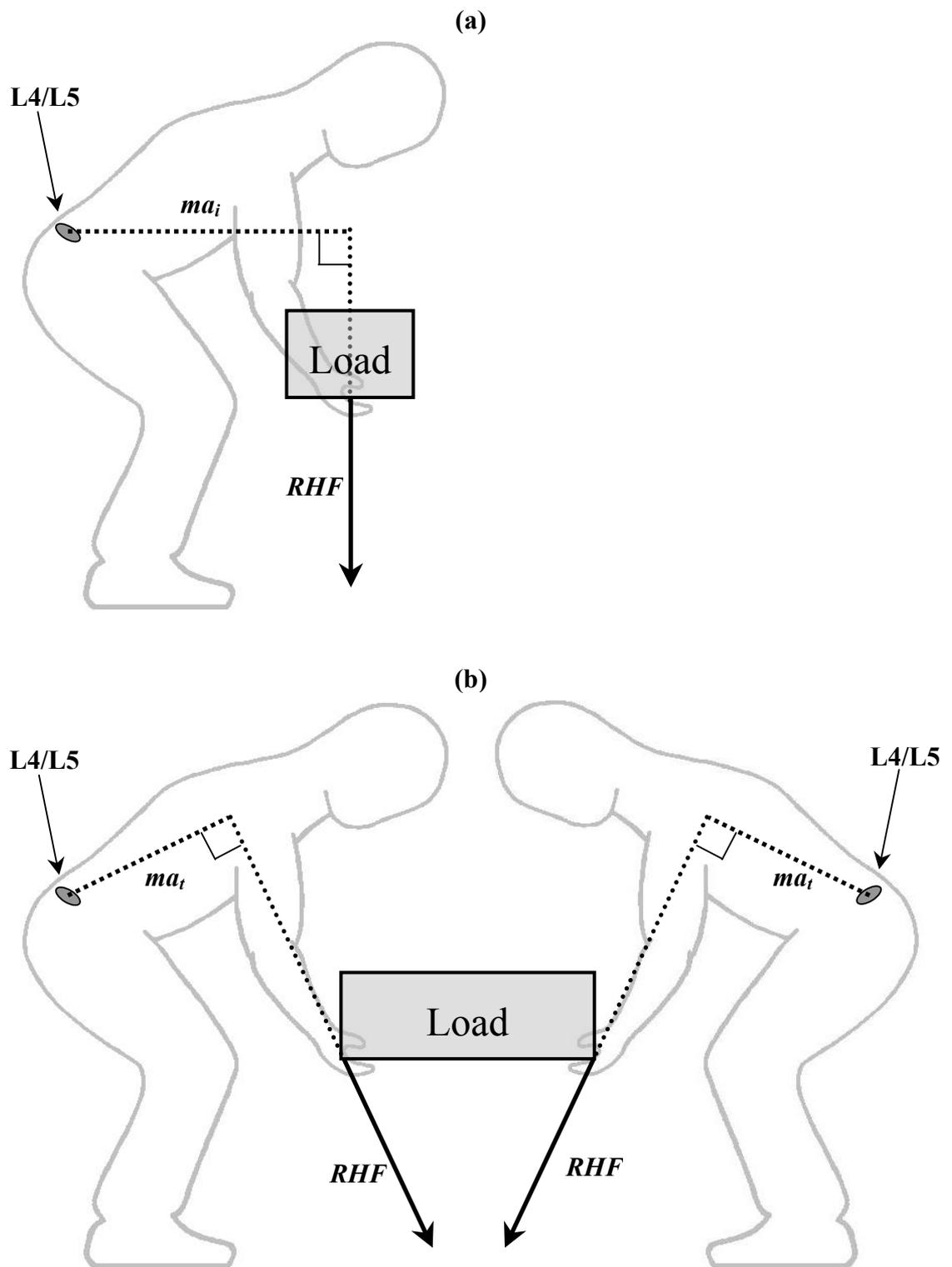


Figure 1.2: Forces and moment arms during: (a) an individual lifting task and (b) a team lifting task. (RHF = Resultant Hand Force, ma = moment arm of the RHF relative to L4/L5 for individual (ma_i) and team lifts (ma_t)). Note $ma_i > ma_t$.

moment arm of the load during a team lift can be less than during an individual lifting task (Figure 1.2). A reduction in the moment arm of the RHF will reduce the net flexion torque produced by the load. As a consequence, less compression force is exerted on the lumbar spine during team lifting tasks compared to during individual lifting tasks. In order to examine the effect of the RHF direction on lumbar spinal load a static model of the lumbar spine was developed, which is presented in Appendix A. Despite possible load relieving benefits no study has measured HHF or its affect on spinal load during team lifting.

1.2 Statement of the problem

Spinal injury is a significant problem in the workforce resulting in large economic and social costs to the community. Of the multi-factorial causes of LBP, lifting has been the most commonly implicated source of spinal injury due to the large spinal loads that often occur. Consequently, manual handling guidelines have been developed by relevant governing bodies of industrialised nations in an attempt to reduce the risks of spinal injury during manual handling and thus reduce the economic cost of LBP in the community. Whilst the guidelines pertaining to manual handling tasks performed by individuals are extensive, there are very few guidelines that apply specifically to manual handling tasks performed by lifting teams.

As previously stated, the Australian Manual Handling: National Code of Practice recommends that team lifting should not be used as the first option in risk control (NOHSC, 1990). While this recommendation may be justified as a conservative risk control strategy, few studies have actually examined the risks associated with team lifting. Indeed, the limited number of studies on team lifting to date have indicated that under certain conditions spinal loads were lower during team than individual lifts, even when the load was twice as heavy in the team lifting tasks (Mital and Motorwala, 1995; Marras et al., 1999).

One reason that spinal loads can be lower in team compared to individual lifting tasks may be because team members are able to exert a horizontal ‘pulling’ force on the load whilst lifting. As a result, reaction torque produced by the load may be lower in team than individual lifting, which will reduce the spinal loads experienced in team lifting tasks. However, up until this point the existence of a pulling force as a spinal load relieving mechanism during team lifting has not been investigated. Additionally, further investigation is needed on the effects of matching team member height and uneven load mass distribution on spinal load during team lifting. Therefore, further investigation on team lifting is justified, so that more specific team lifting guidelines can be developed that are based on scientifically valid empirical evidence.

1.3 Purpose and hypotheses

The general purpose of this study was:

To examine factors that influence spinal loads during two-person team lifting, thereby identifying the spinal injury risks associated with using team lifting as a manual handling strategy.

The specific objectives of the experiments comprising this study were:

- 1) To compare the spinal loads under individual and two-person team lifting conditions and to identify the main biomechanical factors responsible for any observed differences.
- 2) To determine the effect of relative team member height (matched versus unmatched) on lumbar spinal loads during two-person team lifting tasks and to identify the main biomechanical factors responsible for any observed differences.
- 3) To examine the relationship between load mass distribution and the spinal loads incurred by each of the individual team members during two-person team lifting tasks.

The null hypotheses of the experiments comprising this study were that:

- 1) There is no significant difference in the spinal loads experienced during individual versus team lifting tasks with equivalent per person load mass.
- 2) There is no significant difference in the spinal loads experienced by the individual team members of height matched versus height unmatched lifting teams.
- 3) A change in the distribution of the load mass between the individuals of a two-person lifting team results in an equivalent change in the distribution of spinal load between those team members.

1.4 Significance of the study

Considerable financial and social costs are associated with manual handling induced LBP in the workplace (Keyserling, 2000). Given that 53% of all manual handling tasks are performed by teams in some workplaces (Sharp et al., 1997), the assessment of spinal injury risks associated with team lifting is warranted.

Although team lifting is commonly used in the workplace, the current regulations pertaining to its use are rather simplistic. Australian OHS guidelines (NOHSC, 1990) suggest that team lifting should only be used only as a last resort and if team lifting is unavoidable that team members should have a similar height and build. No reference is made to the distribution of the load mass and its relation to the risk of spinal injury. Additionally, the Queensland Department of Industrial Relations recommends avoiding team lifting because the lifting capacity of a team is less than the combined lifting capacity of the individual team members (WHS, 2000). Interestingly, research has both supported (Kowalski, 1988; Lee and Lee, 2001) and opposed (Mital and Motowala, 1995; Marras et al. 1999) this statement. Furthermore, the biomechanical factors that can influence spinal load during team lifting have not yet been identified.

Therefore, this study aims to provide empirical evidence on spinal load during team lifting and thus highlight the relative risks associated with team lifting tasks. In particular, the necessity of team members to be height matched was investigated along with the relative spinal load distribution between team members performing lifting tasks with asymmetrical loads. It is intended that the results from these experiments can be used to further develop the guidelines on safe team lifting practices and thus reduce the number and severity of occupationally related spinal injuries in the workplace.

Chapter 2

Literature Review

The purpose of this chapter is to review the research relevant to team lifting. Firstly, the incidence and costs associated with lower back pain (LBP) are detailed, highlighting the significance of the LBP problem. The multi-factorial causes of LBP are also identified, particularly with respect to manual handling induced spinal injury. Next, research that has examined the structure and function of the lumbar spine is integrated with the mechanisms of spinal injury. Recommendations for safe lifting practices and current occupational health and safety (OHS) guidelines on manual handling and team lifting in particular are also outlined. Finally, the literature review concludes with the research that has specifically investigated team lifting.

2.1 Incidence and costs of low back pain

Lower back pain (LBP) is common in the general population with incidence rates in industrialised countries being estimated at nearly 70% across the lifespan (Andersson, 1981), although these estimates can be as high as 85% in the workforce (Spengler et al., 1986). The high incidence of LBP related illness in the workforce results in equally high financial cost on the economy, resulting in LBP being the most expensive injury with occupational origins (Leamon, 1994).

Chaffin and Park (1973) indicated that for industrialised nations 10 – 30% of all medical claims were due to lumbar spinal injuries. In Australia, worker compensation claims resulting from lumbar spinal injuries amounted to \$AUD1.2 billion in the 1994/95 financial year (NOHSC, 1996). As well as the fiscal costs of workers compensation claims, the reduced productivity associated with absence from work, increased staff turnover and retraining all put added financial strain on the community (NOHSC, 1995).

In the United States of America, the financial costs associated with LBP have been reported to be anywhere from \$US13 – 20 billion annually (Snook and Webster, 1987; Frymoyer, 1992; Nachemson, 1992; Keyserling, 2000), although total annual costs to society have been estimated to be as high as \$US100 billion (NIOSH, 1996). According to the World Health Organisation (WHO, 1999), similar financial costs per capita are experienced by developing countries, with both short term and permanent work disability due to LBP leading to economic losses amounting to as much as 5% of Gross National Product. Although the social costs associated with LBP are more difficult to evaluate than the financial costs, their effects on both individuals and society should not be underestimated (Lancourt and Kettelhut, 1992).

Even though the prognosis for LBP is good, with 80% of LBP cases being alleviated in 2 – 4 weeks (NIOSH, 1997), more than half of those cases experiencing an initial bout of LBP will go on to suffer recurring back pain throughout their lives (Hultman, 1987). In contrast to the majority of individuals, the remaining 20% of LBP sufferers have a chronic pathology that results in 80% of the total financial cost to the community (Valat et al., 1997; NIOSH, 1997). The prognosis for chronic LBP sufferers is not so good, with almost half of those who have not returned to work after six months being transferred onto a long term permanent disabilities pension after twelve months regardless of the post injury intervention strategy employed (Hagen and Thune, 1998). Thus, as chronic LBP sufferers have the worst prognosis with the highest associated costs, the most effective way to reduce these costs in the long term is to implement prevention rather than post-injury management strategies. However, preventing lumbar spinal injury is a difficult task as many factors have been linked either directly or indirectly to LBP.

2.2 Aetiology of lower back pain

In order to examine the causes of LBP, research must: (i) identify factors that are statistically related to LBP via epidemiological studies and (ii) examine those factors experimentally to reveal the underlying mechanisms that lead to LBP.

2.2.1 Epidemiological research examining factors linked to lower back pain

Epidemiological research techniques have been extensively used to identify factors that are related to the incidence of LBP. Some factors that have been specifically examined by epidemiological research for links with LBP include: age, weight, gender, ethnicity, whole body vibration, environmental temperature conditions, physical strength capacity, fatigue, manual handling (lifting, carrying, pushing and pulling), previous LBP, psychosocial factors (e.g. stress, degree of co-worker social support, level of job satisfaction) and even the degree of legal representation. Table 2.1 presents a selected list of some of the epidemiological research that has been conducted with the aim of identifying factors that are related to LBP. The factors that were examined in each study are identified as the independent variable in Table 2.1.

Table 2.1: Summary of epidemiological research examining factors linked to LBP.

Authors	Independent variables	Summary of major results
Bigos et al. (1991)	Various psychosocial factors in the workplace (e.g. job satisfaction).	The perception of employees to the working social environment was a predictive factor in the occurrence of LBP.
Chaffin & Park (1973)	Occupational lifting strength requirements, age, weight, height and previous LBP.	Only the requirement of large occupational lifting strength was significantly correlated with LBP.
Grimmer & Williams (2000)	Age, gender, weight, backpack load, time carrying loads, time sitting and playing sport.	Increased backpack loads, increased carrying duration, increased time sitting and being female all significantly increased the likelihood of reporting LBP.
Hoozemans et al. (1998)	Pushing and pulling tasks.	Increased force requirements during pushing and pulling tasks were related to an increased incidence of LBP.
Ismail et al. (1999)	Vertebral deformities, age and gender.	Vertebral deformities were more prevalent in older females and were significantly correlated with LBP.
Kerr et al. (2001)	A variety of biomechanical and psychosocial factors.	Peak lumbar shear, peak load handled, cumulative disc compression, a physically demanding job, poor workplace social environment, and surprisingly better job satisfaction and co-worker support were all significantly correlated with LBP.
Keyserling (2000)	A variety of biomechanical and psychophysical factors during lifting.	Biomechanical loads on the lumbar spine during lifting were the most predictive factor of LBP.
Kumar (1990)	Cumulative spinal load during lifting.	Cumulative compression and shear forces at the lumbar spine during lifting were significantly correlated with the incidence of LBP.
Leboeuf-Yde (2000)	Body weight.	Body weight has only a weak positive correlation with LBP.
Lings & Leboeuf-Yde (2000)	Whole body vibration.	An environment where whole body vibration exists increases the likelihood of suffering LBP.
Luoma et al. (2000)	Disc degeneration.	The presence of degenerative discs had a significant positive correlation with LBP.
Neuman et al. (2001)	Spinal load and kinematic variables.	Spinal load during lifting, lifting duration, average shift load and time spend with a flexed trunk were all positively correlated with LBP.
Norman et al. (1998)	Spinal load.	Acute and cumulative compressive and shear forces on the lumbar spine were positively correlated with LBP.
Pienimaki (2002)	Environmental temperature.	Exposure to cold working conditions significantly increased the incidence of LBP.
Tafazzoli & Lamontagne (1996)	Hamstring stiffness.	People with increased hamstring stiffness were more likely to suffer LBP than those with more relaxed hamstrings.
Tait & Chibnall (2001)	A variety of psychosocial variables.	Ethnicity and degree of legal representation were positively correlated with the duration of LBP episodes.
Wilder & Pope (1996)	Whole body vibration.	Whole body vibration significantly increased the incidences of LBP.

As shown by Table 2.1, there are many factors that have various degrees of correlation with the incidence of LBP. As a result, LBP is generally referred to as having a multi-factorial aetiology (Nachemson, 1992; Keyserling, 2000). Some of the most commonly investigated factors examined by epidemiological research are psychosocial variables such as stress and worker satisfaction (Svensson and Andersson, 1989). Yet, despite the large number of studies that have been conducted, no specific psychosocial risk factors have been consistently associated with LBP (Ferguson and Marras, 1997). Of all the factors that have been examined epidemiologically, manual handling and specifically lifting has been the most commonly implicated factor causing LBP (Magora, 1973; Andersson, 1981; Kelsey, 1984; Bigos et al., 1986; Burdorf and Sorock, 1997). In particular, excessive spinal load (acute or cumulative) during lifting has been shown to be the factor with the strongest correlation to LBP (Norman et al. 1998; Keyserling, 2000). This link between LBP and excessive spinal load on the lumbar spine during lifting suggests that LBP originates from damage to the biological structures of the lumbar spine (McGill, 1997).

The advantage of the epidemiological research approach is that the relationship between the effector (e.g. excessive spinal load) and the symptom (i.e. LBP) can be established. However, although the epidemiological approach can specify which factors are correlated with LBP, it does not reveal the underlying mechanisms that cause spinal injury nor can it help identify under what conditions are these factors most likely to occur.

2.2.2 Experimental research examining the relationship between manual handling and spinal injury

Experimental research is required to examine the mechanisms underlying the causes of spinal injury during manual handling. In order to identify the conditions that cause spinal injury during manual handling, experimental studies have been conducted using psychophysical, physiological and biomechanical approaches.

Psychophysical research methods typically utilise estimates of the physical effort required to perform a particular task over a full working day, based on the discrete performance of that task in the laboratory. Comparisons are then made between the psychological self based evaluations of one manual handling task compared to other tasks. Results of selected research projects that have been conducted using the psychophysical approach are summarised in Table 2.2.

The foundation of the psychophysical methodological framework is that the risk of injury increases as the subject finds the task more physically demanding. Although this type of investigation can identify tasks that may incur greater risks of injury when the task demands are greater, the mechanism causing the increased risk is the increased mechanical loading of musculoskeletal structures (Dempsey, 1998). Thus, psychophysical studies only give an indication of the risk of spinal injury by indirectly estimating spinal load. Fernandez et al. (1991) indicated that during 25-minute laboratory sessions maximum acceptable weight limits for an 8-hour working day were over estimated by nearly 20%. Furthermore, the subjective evaluation estimates incorporated by psychophysical studies have been shown to change over even short periods of time (Wu and Chen, 2003).

Table 2.2: Summary of psychophysical research investigating the causes of manual handling induced spinal injury.

Authors	Independent variables	Summary of major results
Barker & Atha (1994)	Lifting training.	Lifting training did not reduce the effort required to lift loads.
Chen (2003)	Tightness of abdominal belts whilst lifting.	The tightness of the abdominal belt significantly increased the maximum acceptable weight limit.
Chung & Wang (2001)	Carrying loads up stairs.	Stair climbing significantly influenced both maximum weight lifting capacity and oxygen consumption.
Ciriello (2001)	Box size and vertical lowering height.	Box size and the vertical height did not significantly affect the maximum acceptable weight limit.
Ciriello et al. (2001)	Pushing tasks on floor surfaces with high and low coefficients of friction.	The maximum acceptable load was significantly greater for pushing tasks performed on floor surfaces with a higher coefficient of friction.
Drury et al. (1989)	Lifting boxes with a variety of handle positions and no handles.	The provision of box handles lowered both RPE and heart rate, although the optimum position was dependent on the weight of the load.
Gallagher (1991)	Various vertical lift distances with stooped, kneeling, symmetrical and asymmetrical postures.	Psychophysical lifting capacity, heart rate and oxygen consumption were all the lowest whilst lifting with a symmetrical posture to the lowest shelf whilst kneeling.
Haslam et al. (2002)	Pushing and pulling on floor surfaces high and low coefficients of friction.	No significant difference was found between the maximum acceptable load during pushing and pulling tasks performed on floor surfaces with high or low coefficients of friction.
Lee et al. (1992)	Back versus side lifting techniques.	The maximum acceptable weight limit was significantly greater for back lifts compared to side lifts.
Lee & Lee (2001)	Lifting teams of matched versus unmatched heights	Greater lifting capacity was measured when the team members had matched standing height.
Mital (1992)	Symmetrical versus asymmetrical lifting tasks.	The maximum acceptable weight limits were nearly identical for symmetrical and asymmetrical lifting tasks.
Parikh et al. (1997)	Asymmetrical lifting and lowering tasks.	Maximum acceptable lifting capacity significantly decreased as the vertical height of the asymmetrical lifting task increased.
Snook (1971)	Age.	Lifting capacity was negatively correlated with age.
Stalhammar et al. (1996)	Lifting frequency and parcels with and without handles.	Acceptable lifting frequency was significantly higher for parcels with handles than for those without handles.
Straker & Duncan (2000)	Stoop versus squat lifting.	Lifting capacity and RPE were significantly greater for squat than stoop lifting.
Wu (1997)	Box size and lifting frequency.	Increases in box size and lifting frequency induced significant decreases in the maximum acceptable weight limit during lifting.
Zhu & Zhang (1990)	Lifting technique (stoop, squat and freestyle), lifting frequency and anthropometric characteristics.	A freestyle lifting technique had the highest acceptable workload followed by stoop then squat lifting. A decrease in lifting frequency and an increase in anthropometric body size were significantly correlated with an increased maximum acceptable workload.

Physiological research techniques measure factors such as oxygen consumption and heart rate to identify tasks that incur greater risk of spinal injury during manual handling. The assumption of the physiological paradigm is that the greater the physiological stress on the body the greater the risk of spinal injury during manual handling tasks (Åstrand and Rodahl, 1986). The theoretical basis of the physiological model of spinal injury is that increased physiological stress can lead to local or general muscular fatigue, which may cause the individual to perform the task in a manner that increases the risk of injury due to overexertion (Asfour et al., 1988; Hoozemans et al., 1998). Table 2.3 presents summaries from selected physiological research on manual handling.

Physiological research into manual handling has primarily focused on investigating lifting technique. Using physiological methods to study lifting techniques, it has generally been concluded that squat lifting incurs a higher physiological cost than lifting with a stooped technique (Welbergen et al., 1991; Duplessis et al., 1998). Although the physiological approach can indicate which lifting technique incurs the greatest physiological cost, the link to LBP is dependent on the fatigue of the muscular system. Therefore, the usefulness of the physiological approach when examining LBP caused by discrete lifting tasks is limited. Furthermore, although physiological changes can be used to identify tasks with increased risk of spinal injury, it is the resulting kinematic changes in the performance of the task and the increased spinal load that eventually causes the spinal injury itself.

Both psychophysical and physiological research techniques have been used to identify tasks that increase the risk of spinal injury. However, both of these techniques are indirect estimates of spinal load, which if excessive is the actual cause of spinal injury.

A more direct method of measuring spinal load is via biomechanical techniques. Table 2.4 summarises the major findings from some experimental research that has investigated manual handling using the biomechanical approach.

Table 2.3: Summary of physiological research investigating the causes of manual handling induced spinal injury.

Authors	Independent variables	Summary of major results
Beynon et al. (2000)	Rest schedules.	Altering the frequency of the rests periods did not significantly affect spinal shrinkage or physiological fatigue.
Birch & Reilly (1999)	Lifting during various phases of the menstrual cycle.	The menstrual cycle had no significant effect on any physiological measure during lifting.
Chung & Wang (2001)	Carrying loads up stairs and gender.	Carrying loads up stairs caused a significant increase in oxygen consumption. Oxygen consumption was also significantly greater for females as compared to males whilst carrying equivalent loads.
Duplessis et al. (1998)	Stoop and squat lifts with and without back belt supports.	Whilst the squat technique increased oxygen consumption by 23% compared to stoop lifting the back belt had no significant effect.
Dury et al. (1989)	Load weight and asymmetrical hand positions whilst lifting.	The use of handles and reducing the load weight significantly decreased heart rate. However, the use of asymmetrical compared to symmetrical hand positions lowered heart rate when the load mass was low and increased it when the load mass was high.
Garg & Herrin (1980)	Stoop versus squat lifting techniques.	Squat lifting incurred significantly higher oxygen consumption than stoop lifting.
Garg & Saxena (1979)	Lifting frequency.	Lifting frequency was significantly correlated with metabolic consumption rates.
Laursen et al. (2000)	Uphill, level and downhill carrying of loads.	During carrying tasks, oxygen consumption is linearly correlated with the slope of the ground.
Lee & Chen (1995)	Seated and standing lifting tasks.	Tasks performed whilst sitting incurred higher physiological cost than during standing.
Mital et al. (1994)	High lifting frequency.	High lifting frequency (16 reps/min) tasks performed by experienced manual handlers incurred physiological costs (i.e. heat rate and oxygen consumption) that were well above physiological design criteria.
Ruckert et al. (1992)	Load mass, location and number of repetitions during lifting.	Increases in the: weight of the load, horizontal distance the load is from the person and number of tasks required all significantly increased heater rate and RPE.
Smith & Jiang (1984)	The mass and ‘fullness’ of a bag during lifting.	Although the mass of the bag significantly increased the physiological stress on the body, the amount of material within the bag (i.e. the ‘fullness’) had no significant effect.

Table 2.4: Summary of biomechanical research investigating the causes of manual handling induced spinal injury.

Authors	Independent variables	Summary of major results
Anderson & Chaffin (1986)	Lifting techniques.	A squat technique with a straddled foot stance yielded the lowest spinal load.
Authier et al. (1996)	Novices and expert lifters.	The kinematics of the novices was significantly different from that of the expert lifters.
Buhr & Chaffin (1997)	Leg strength.	Decreases in leg strength causes kinematic changes in lifting technique that can increase spinal load.
Burgess-Limerick & Abernethy (1997)	Load mass.	Self selected stoop and semi-squat techniques were used for light loads, whilst only semi-squat techniques were observed for heavier loads.
Bush-Joseph et al. (1988)	Lifting speed.	Spinal load increased proportional to increases in lifting speed.
Callaghan & McGill (2001b)	Prolonged sitting.	Prolonged sitting resulted in increased EMG activity of the lumbar musculature that may result in fatigue and increased risk of LBP.
Cholewicki & McGill (1995)	Various manual handling tasks.	Spinal joint stability increases with increasing task demands. LBP risk is minimised when the trade-off between joint instability and spinal loading are optimal.
Danz & Ayoub (1992)	Speed, frequency and load mass.	Very few lifters performed lifts smoothly, which increased spinal load.
de Looze et al. (1999)	Load mass during lifting.	Increased abdominal activity during lifting increased the load on the lumbar spine.
de Looze et al. (1998)	Asymmetrical leg position during lifting.	An asymmetric leg position did not significantly reduce spinal loads or muscle activity.
Dolan & Adams (1998)	Repetitive lifting.	Repetitive lifting significantly fatigues the spinal extensors and increases the bending moment acting on the spine.
Dolan et al. (1994)	Lifting technique, load mass, speed, horizontal distance and asymmetry.	Stooped lifting, increased speed and twisting all significantly increased spinal load.
Hooper et al. (1998)	Asymmetric lifting tasks.	Asymmetrical lifting tasks caused similar asymmetrical loading patterns in the lumbar spine.
Kim & Chung (1995)	Posture, weight and frequency during lifting.	Lumbar muscles became fatigued faster for light load-high frequency tasks than for heavy load-low frequency tasks.
Kumar et al. (1998)	Flexion and rotation tasks.	Trunk strength is reduced in a flexed and rotated position. Thus, back injury may be precipitated by a combination of flexion and twisting movements.
Larivière et al. (2000)	Patients with and without pre-existing LBP.	Pre-existing LBP may increase the risk of spinal injury during lifting due to increased trunk flexion.
Lavender et al. (2003)	Initial vertical lift height, load mass and lifting speed.	Spinal load significantly increased when, lifting from a lower height, load mass was increased and with increased lifting speed.
Lee & Lee (2002)	Unstable loads during lifting.	Muscle activity of the trunk musculature increased when sudden perturbations were applied to the load.

Table 2.4: (Continued).

Authors	Independent variables	Summary of major results
Leskinen et al. (1992)	Self-paced and force-paced lifting.	No significant difference in spinal load occurred between self-paced and force-paced lifting.
Marras et al. (1999)	Box size, shape and handles during lifting.	The inclusion of handles on the box significantly reduced spinal load.
McGill et al. (2000)	Lumbar lordosis.	Anterior shear spinal load is significantly reduced when lifting with a lordotic rather than a flexed lumbar curvature.
Oddsson et al. (1999)	Perturbations during lifting.	Perturbations during lifting resulted in significant increases in EMG activity of the lumbar muscles.
Schultz et al. (1982)	Trunk flexion and twisting whilst holding a load.	Spinal load was significantly correlated with the amount of trunk flexion and rotation.
van der Burg et al. (2000)	Lifting an unexpectedly heavy object.	Lifting an unexpectedly heavy object did not significantly increase spinal load or reduce balance.
van Dieën et al. (1996)	Induced fatigue during lifting.	Spinal load and kinematic coordination were not unduly affected by muscle fatigue.
van Dieën et al. (1999)	Stoop versus squat lifting.	There was no significant difference in the biomechanical risks associated with using either a squat or a stoop lifting technique.

Because spinal injury leading to LBP can only occur as a result of excessive mechanical load on the musculoskeletal structures of the spine (McGill, 1997), a great deal of research has been directed towards quantifying spinal loads during a variety of activities (Schultz et al., 1982; Bush-Joseph et al., 1988; van Dieën et al., 1999; van der Burg et al., 2000). Of all the activities that may cause LBP, manual handling has been the most commonly implicated factor (McGill, 1997; Dolan and Adams, 1998). Lifting in particular has been singled out as an activity that can experience large spinal load, especially when lifting heavy objects (NIOSH, 1997). Primarily it is the excessive spinal load during manual handling tasks that leads to spinal injury and eventually LBP. As well as directly initiating spinal injury, spinal load can also affect inter-vertebral stability that may also lead to spinal injury in certain circumstances (Cholewicki and McGill, 1995; Cholewicki et al., 2000).

Manual handling tasks involving; large loads, awkward lifting postures, rotational movements, high frequency, long duration and many repetitions incur the highest spinal loads. However, in order to understand how spinal loading can cause spinal injury an appreciation of the mechanics of the lumbar spine is required.

2.3 Mechanics of the lumbar spine

2.3.1 Structure and function of the lumbar spine

The anteriorly convex lumbar spine is comprised of the vertebrae (L1 to L5 plus the sacrum), inter-vertebral discs located within the joints L1/L2 through to L5/S1, the connective ligamentous tissues surrounding these joints and the tendinous and muscular structures that help stabilise and control movement within those joints. Each of these biological tissues plays an important role in enabling the lumbar spine to withstand the mechanical stress placed upon it, whilst enabling a functionally sufficient range of motion.

The five lumbar vertebrae (L1, L2, L3, L4 and L5) are comprised of a relatively large and oval shaped vertebral body, between which the inter-vertebral discs reside. Directly posterior to the vertebral bodies is the vertebral foramen, through which the spinal cord passes. Surrounding the vertebral foramen is the vertebral arch from which the articular processes, the two lateral transverse processes and the superior and inferior posterior spinous process protrude (Figure 2.1).

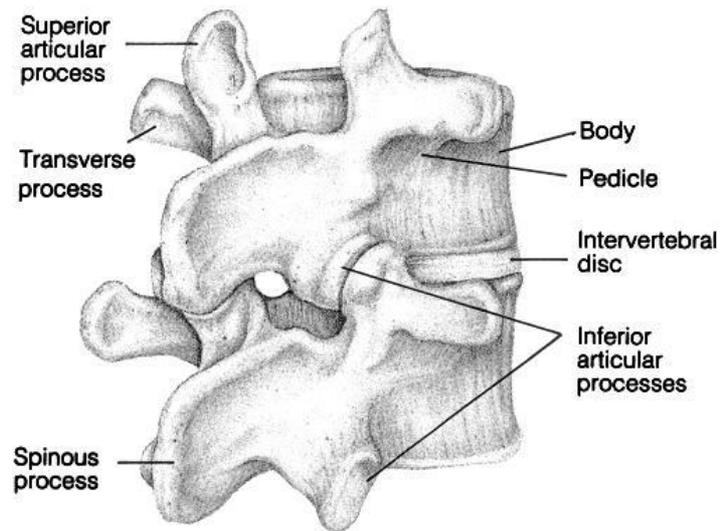


Figure 2.1: Anatomical structure of the lumbar vertebrae and inter-vertebral disc (Marieb, 1995).

As the lumbar spine is the most inferior section of the spine it bears more weight than either the thoracic or cervical sections, and thus the vertebral bodies are larger to compensate for the larger compressive forces. The transverse and spinous process are attachment points for ligaments and tendons that provide the stability and movement within the lumbar joints. However, unlike the thoracic vertebra, the ribs do not attach to the transverse processes thus enabling more mobility in the lumbar spine. The mobility within the lumbar joints is primarily a flexion/extension movement (approximately 14° – 20° per segment) due to the sagittal alignment of the articulating processes, although lateral flexion/extension (approximately 3° – 8° per segment) and rotation in the transverse plane (approximately 2° per segment) can still occur (Rasch, 1989).

The inter-vertebral discs account for about 25% of the length of the vertebral column and are made up of two parts, the nucleus pulposus and the annulus fibrosus (Figure 2.2). Shock-absorbing properties of the discs are provided by the nucleus pulposus, which is a flattened spherical structure residing in the centre of the disc that counteracts the compressive forces within the lumbar spine. The annulus fibrosus is attached to

successive vertebra and is made up of concentric rings of fibrocartilage tissue around the nucleus that are sequentially arranged in oblique, transverse and longitudinal directions. Consequently, some of the fibrous material of the annulus will always be in tension when the joint is in a non-neutral position, which limits the range of motion to help prevent injury.

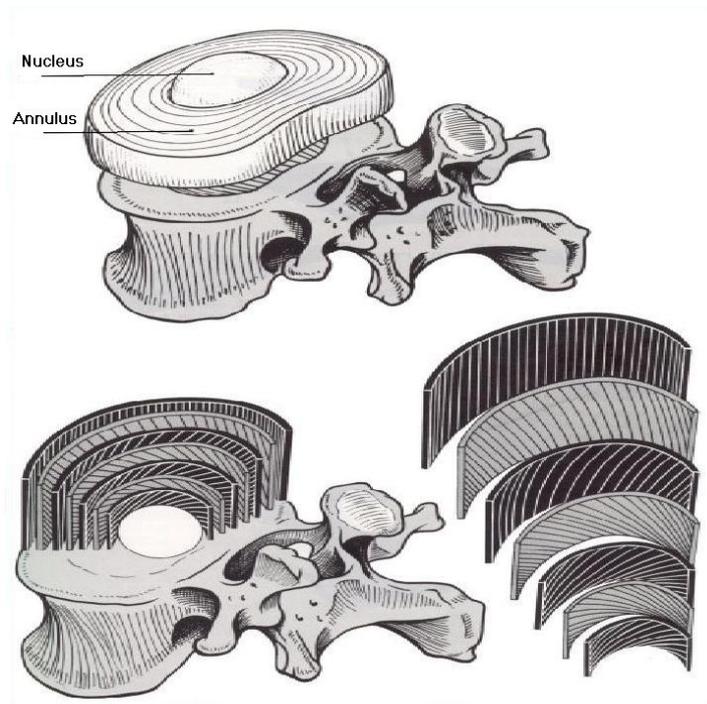


Figure 2.2: Anatomical structure of the inter-vertebral disc (Kapandji, 1974).

There are numerous ligaments attached between the adjacent vertebra of the lumbar spine (Figure 2.3), which act to stabilise the spinal joints. Particularly, during flexion/extension movements (Dolan and Adams, 1994). Excessive extension is limited by the anterior longitudinal ligament, whilst excessive flexion is arrested by the posterior longitudinal ligament, ligamentum flavum, the interspinous ligaments and the supraspinous ligament. The alignment of the interspinous ligaments also provides resistance to posterior shear and torsional force between adjacent spinous vertebrae. However, in a fully flexed position the alignment and tension of the interspinous ligament relative to the spinal column can actually generate an anterior shear force (Heylings, 1978; McGill, 1988; McGill et al., 1998).

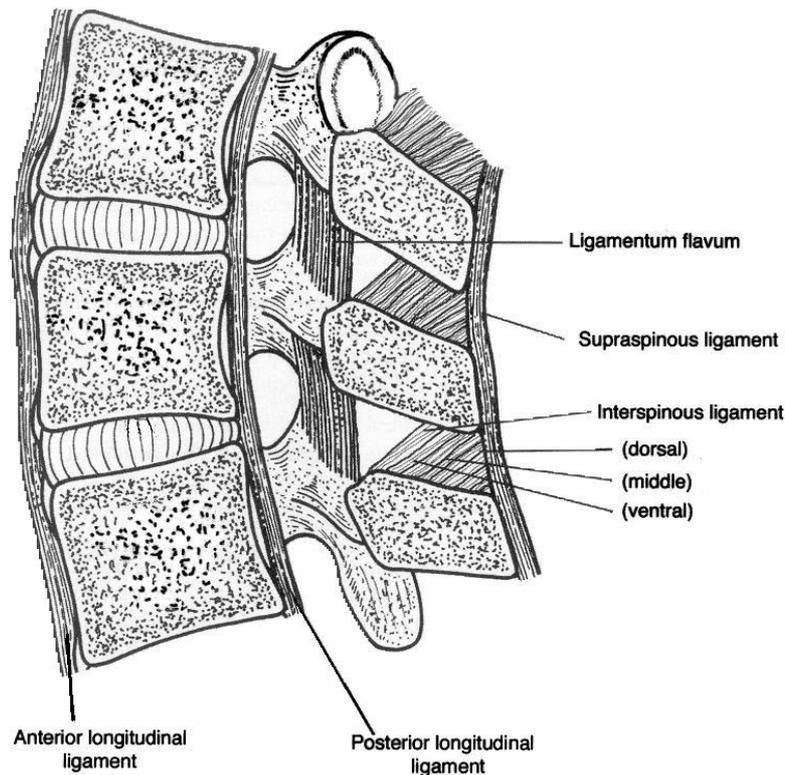


Figure 2.3: Anatomical structure of the spinal ligaments (McGill, 2002).

Although the anatomy of the lumbar musculature is very complex, the major muscles controlling the movement and stability of the lumbar spine are: latissimus dorsi, quadratus lumborum, rotatores, intertransversarius, multifidus, interspinales, psoas minor, psoas major, iliacus, the erector spinae group and the abdominal group (Bogduk, 1980; Bogduk et al., 1998; Stokes and Gardner-Morse, 1999). Many of these muscles such as intertransversarius and multifidus act to stabilise the spine by controlling the fine inter-vertebral moments that keep the spine properly aligned (McGill and Norman, 1987a; Cholewicki et al., 1996). As the movement within the lumbar spine is primarily flexion/extension, the erector spinae and abdominal groups of muscles can be considered the prime movers of the lumbar spine. The group of muscles commonly referred to as the erector spinae are actually four muscles: longissimus thoracis, iliocostalis lumborum, sacrospinalis and multifidus (Bogduk, 1980; Bogduk et al., 1992). Apart from multifidus, whose main function is to produce small vertebral adjustments (Donisch and Basmajian, 1972), the erector spinae group generates the primary extension torque that keeps the body upright and enables objects to be lifted

and carried. As well as extending the lumbar spine, the lumbar extensors can generate a posterior shear force due to the oblique line of action of the erector spinae group when the lumbar spine has a lordotic curvature (Potvin et al., 1991; McGill et al., 2000).

Opposing the extension torque produced by the erector spine is the flexion torque produced primarily by rectus abdominis and the internal and external obliques. The fourth muscle of the abdominal group, transversus abdominis, is more likely to be involved with spinal stability and intra-abdominal pressure than lumbar movement, due to the transverse orientation of the muscle fibres (McGill and Sharratt, 1990; Cholewicki et al., 1999; Cholewicki and VanVliet, 2002).

2.3.2 Mechanics of injury to the lumbar spine

LBP is the result of direct or indirect activation of one or more of the pain receptors that reside in the lumbar spine. Upon activation of these receptors pain is registered via the release of pain mediating chemicals such as enkephalines and serotonin. Additionally, the production of these pain mediating chemicals can be influenced by psychological states, which explains why pain is perceived more strongly when a person is depressed, exhausted, or disgruntled (Nachemson, 1992). Even though the predisposition to experience pain may vary, occupationally related LBP is caused by musculoskeletal injury that is biomechanical in nature (McGill, 1997; Kumar, 2001) and occurs due to either acute or cumulative force beyond the structural limits of the spinal tissues.

The most obvious cause of spinal injury occurs when a discreet load is applied to the lumbar spine that is beyond its absolute tolerance limit. Such an incident may often occur during a 'slip and fall' accident where the person lands on a hard surface with their coccyx (Manning and Shannon, 1981; Manning et al., 1984; Omino and Hayashi, 1992). The resulting sharp axial load applied to a flexed spine may produce a compression force beyond the structural limits of the lumbar spine (Figure 2.4a), which can cause fracture of the vertebral end plate (Troup et al., 1981; Bigos et al., 1986;

McGill, 1997). Conversely, cyclic loading of the lumbar spine below the structural limit of the spinal tissues may not produce spinal injury on the first cycle, but over time the ability of the tissues to withstand the applied load may decrease resulting in prospective spinal injury (Figure 2.4b). Similarly, prolonged flexion of the lumbar spine may induce tissue creep that decreases the tolerance of the spinal tissues, thus making them more susceptible to spinal injury (Figure 2.4c). Although excessive acute force can cause spinal injury, a more common cause is as a result of cumulative load (Figures 2.4b and 2.4c) applied over a prolonged duration (Kumar, 1990; McGill, 1997; Norman et al., 1998).

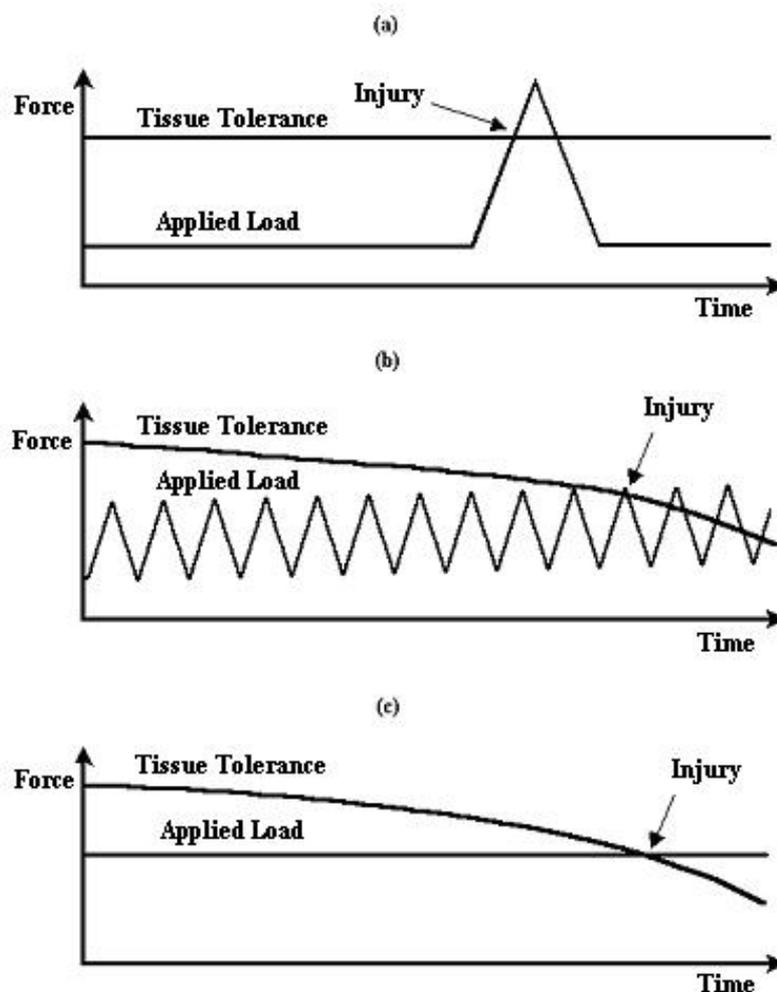


Figure 2.4: Mechanisms of spinal injury via: (a) acute and excessive loading, (b) cyclically applied load and (c) an extended period at the end-range of motion (adapted from McGill, 1997).

Specific types of spinal injury that may occur due to acute or cumulative loading include: spondylolysis, spondylolisthesis, sciatica, stenosis, idiopathic LBP and end plate fractures of the vertebral body, each of which is described below.

Stress fractures of the pars interarticularis (spondylolysis) occur particularly in young athletes who perform repetitive hyper- flexion/extension and rotation movements of the lumbar spine (Hardcastle et al., 1992; Burnett et al., 1996). If the stress fractures are not treated (generally with rest) and the cause of the injury continues a complete fracture of the vertebral arch can ensue, resulting in the ‘slip’ of one vertebral body over another (spondylolisthesis). Although this condition is rare, patients with spondylolisthesis have an increased incidence of LBP attacks, though it is not often chronically disabling (Saraste, 1987).

Sciatica is caused by pressure placed upon one or more of the nerve roots, usually by a herniated disc, resulting in pain originating in the lower back and radiating down the lower extremities. A herniated or prolapse disc occurs when the nucleus of the inter-vertebral disc pushes through a tear in the annulus (usually posteriorly), and consequently can protrude into the vertebral foramen and impinge upon the spinal cord. If tears are already present in the annulus a three-step mechanism of prolonged flexion, axial compression and rapid extension can significantly increase the risk of posterior disc prolapse (Callaghan and McGill, 2001a). Prolonged flexion of the lumbar spine will cause the nucleus to move posteriorly within the inter-vertebral disc; subsequent axial compression can then impinge upon the nucleus driving part of the nucleus posteriorly through the annulus during a rapid extension of the lumbar spine (Figure 2.5). Disc prolapse is quite rare and can often be asymptomatic (McGill, 1997). However, the recovery rate from prolapse disc related LBP symptoms are very slow and long term chronic pain often follows regardless of the treatment strategy (Andersson et al., 1983).

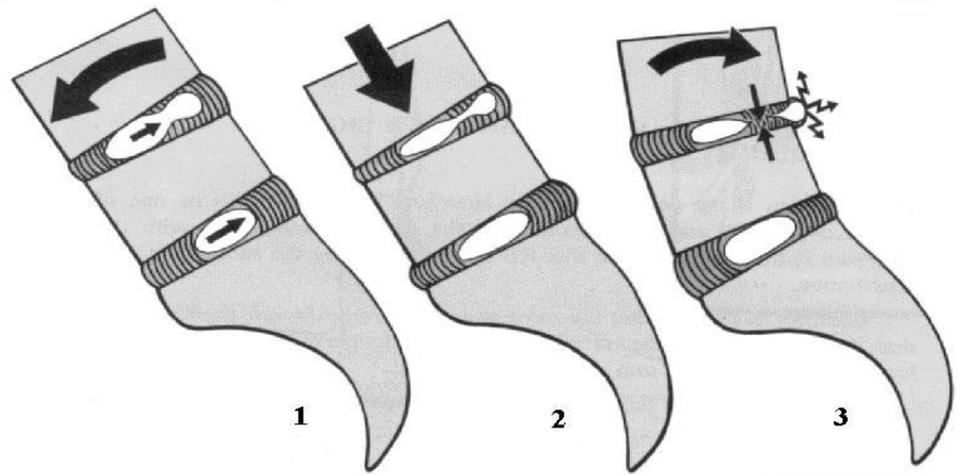


Figure 2.5: Three sequential mechanistic stages of disc prolapse: (1) prolonged flexion, (2) axial loading and (3) rapid extension (Kapandji, 1974).

Idiopathic or non-specific pain is the most common cause of LBP (Nachemson, 1992), with up to 85% of lumbar spinal pain of unknown aetiology (McGill, 2002). However, cumulative loading that results in micro damage to the muscles and/or ligaments is the most likely cause of idiopathic spinal injury (McGill, 1997). Although the cause of idiopathic pain is often difficult to ascertain the prognosis is excellent with over 90% of sufferers returning to work within six weeks (Frymoyer, 1988).

Of all mechanical forces imposed upon the lumbar spine excessive compressive force is the one that is most often linked to spinal injury (Brinckmann et al., 1989; McGill, 1997; Marras, 2000). Excessive compressive force can fracture the vertebral end plates that are situated on the superior and inferior surfaces of the vertebral bodies (Fyhrie and Schaffler, 1992), which can lead to episodes of LBP (Yoganandan et al., 1994). Furthermore, Brinckmann et al. (1989) and Callaghan and McGill (2001b) indicated that the vertebral end plates were the first tissues to be damaged during excessive spinal compression. An end plate fracture results in the localised collapse of the trabecular bone of the vertebral body (Aggrawall et al., 1979; Gunning et al., 2001), which upon healing form Schmorl's nodes that can indicate the amount of damage that has occurred (Vernon-Roberts and Pirie, 1973).

Although compression force is most commonly related to spinal injury, shear force can also impose significant risks to the lumbar spine. Cripton et al. (1995) reported that shearing forces above 2000 N on the lumbar spine pose a risk of injury to the articulating processes between successive vertebrae. Vertebral end plate avulsions have also been known to occur during excessive and repetitive shear loading (McGill, 2002).

Despite the known link between excessive biomechanical force and spinal injury, many other factors must be taken into consideration when attempting to minimise the risk of lumbar spinal injury. For example, the stiffness of the lumbar spine is increased after periods of non load bearing activity due to the flow of fluid back into the inter-vertebral disc, resulting in more tensile stress placed upon the inter-vertebral ligaments and thus increasing the risk of injury (Green et al., 2002; Rozenberg et al., 2002). Other factors such as age (Buckwalter, 1995), prolonged exposure to cold environments (Kakosy, 1989) or whole body vibration (Seidel et al., 1986; Pope et al., 1999; Seidel et al., 2001) can also reduce the ability of spinal tissues to adequately cope with the loads placed upon them by reducing the tissue tolerance limits (McGill, 1997). Thus, to reduce the risk of spinal injury the force applied to the spine should be minimised and/or the tissue tolerance limits should be maximised.

2.3.3 Recommendations for a safe lifting technique based on biomechanical research

The most common cause of LBP is due to manual handling induced spinal injury occurring as a result of excessive loading of the lumbar spine beyond the capabilities of the spinal tissues (McGill, 1997; Viikari-Juntura, 1997; Norman et al., 1998; Keyserling, 2000). Therefore, the risk of spinal injury can be minimised by reducing the loads on the lumbar spine and/or avoiding environmental or mechanical conditions that decrease the ability of the lumbar spinal tissues to deal with mechanical loads. Although it is difficult to recommend a lifting technique that is appropriate for all lifting tasks

(Hsaing, et al., 1997), based upon a knowledge of the functional anatomy of the lumbar spine and research that has investigated spinal load during manual handling some general recommendations can be made with regard to lifting technique. An appropriate lifting technique is designed to reduce the risk of injury during manual handling by increasing the safety margin between spinal load and spinal tissue tolerance.

1) *Reduce the mass of the load being handled.*

One of the most effective ways to reduce spinal load is to reduce the mass of the load that is being handled. Numerous studies have reported a direct association between the mass of the load being handled and the spinal load experienced at the lumbar spine (Norman et al., 1998; Chaffin et al., 1999; Lavender et al., 2003). However, in industrial situations it is often impossible or impractical to reduce the mass of the load (e.g. lifting a person). Thus, in situations where an excessive load mass must be handled and mechanical aids are unavailable particular attention must be focussed on the manual handling techniques that can decrease the risk of spinal injury.

2) *Avoid end of range motion and maintain a neutral lumbar lordosis.*

Full flexion of the lumbar spine, particularly when lifting heavy objects, increases the risk of injury to the passive structures of the lumbar spine. During full flexion of the lumbar spine the posteriorly located lumbar ligaments (posterior longitudinal ligament, ligamentum flavum, interspinous ligaments and supraspinous ligament) are placed in tension, which can result in myoelectric silence of the lumbar extensors (McGill and Kippers, 1994). The stress placed upon the posterior lumbar ligaments during full spinal flexion combined with the lack of stabilising muscular activity places these ligaments at considerable risk of injury, particularly if unexpected perturbations occur during lifting (Radebold et al., 2000; Lee and Lee, 2002). As well as avoiding excessive stress on the lumbar ligaments, maintaining a neutral lumbar lordosis whilst performing a manual handling task will, evenly distribute the compression force across the lumbar discs (Hsiang et al., 1988), minimise anterior shear forces due to the posteriorly oblique

alignment of the lumbar extensors (Potvin et al., 1991; McGill, 1996; McGill et al. 2000) and eliminate the risk of disc herniation (Callaghan and McGill, 2001; McGill, 2002).

3) *Reduce the moment arm of the reaction force at the hands.*

Reducing the moment arm of the reaction force at the hands reduces tissue forces required to support the reaction torque at the lumbar spine, thus reducing the lumbar compression force. During lifting the most common instruction given to reduce the moment arm of the load is to hold the load as close to the torso as possible (NOHSC, 1990; Waters et al., 1993; WHS, 1999). This instruction is an effective means of reducing spinal load as the reaction force at the hands is primarily in a downward direction. However, in situations where the reaction force at the hands is often not vertical (e.g. pushing and pulling tasks) the best way to reduce spinal load is to ensure that the ‘transmissible’ hand reaction force vector is directed as close to the lumbar spine as possible (McGill, 2002).

4) *Avoid generating twisting torque at the lumbar spine.*

To generate a twisting torque at the lumbar spine, such as that required to start a ‘pull-start’ lawn mower, muscle co-activation from a variety of trunk muscles especially the internal and external obliques is required. This muscular co-contraction increases the compressive force experienced by the lumbar spine and thus increases the risk of spinal injury. McGill (1997), has reported that supporting a twisting torque of only 50 Nm can generate compression forces at the lumbar spine of up to 3000 N, which alone approaches the safe Action Limit (3433 N) recommended by NIOSH (1981). Instead of twisting the truck during rotational manual handling movements, pivoting on the balls of the feet will incur lower lumbar spinal load (Gagnon et al., 1993), and thus lower the risk of spinal injury.

5) *Avoid lifting after periods of prolonged flexion.*

Prolonged periods of spinal flexion causes both ligamentous creep and a posterior relocation of the nucleus within the inter-vertebral disc (Jackson et al., 2001; Callaghan and McGill, 2001a). The former decreases the tissue tolerance of the ligamentous tissues (McGill, 1997), whilst the latter increases the risk of a herniated vertebral disc particularly if followed by a strenuous lifting task (Callaghan and McGill, 2001a). Furthermore, McGill and Brown (1992) reported that even after standing for half an hour following a period of prolonged extreme flexion the spinal tissue maintained some residual laxity. Thus, tasks that require extension of the lumbar spine should not be performed after a period of prolonged full lumbar vertebral flexion.

6) *Avoid excessive intra-abdominal pressure and co-contraction.*

Intra-abdominal pressure may produce an extension torque (Morris et al., 1961; Thomson, 1988) that has a load alleviating effect on the lumbar spine. However, current research (McGill and Norman, 1986; Daggfeldt and Thorstensson, 1997) has indicated that the co-contraction of the trunk muscles that accompanies increases in intra-abdominal pressure during lifting, increases the flexion torque and compression force at the lumbar spine to a greater degree than any load alleviating effect that intra-abdominal pressure may produce. Furthermore, intra-abdominal pressure can substantially increase both systolic and diastolic blood pressure (MacDougall et al., 1992) and compromise the return of venous blood (Mantysaari et al., 1984). Conversely, Cholewicki et al. (1999) have found that moderated increases in intra-abdominal pressure can be beneficial during very strenuous lifting tasks by increasing spinal stability. Thus, only a moderate co-contraction of the lumbar musculature is recommended during lifting, which increases spinal stability without drastically increasing spinal load.

7) *Lift in a smooth and controlled manner or use a momentum strategy.*

In an attempt to minimise the inertial force experienced at the hands during lifting, OHS guidelines often recommend that lifting tasks be performed in a slow and controlled

manner (WHS, 1999; NOHSC, 1990). Whilst this lifting technique is appropriate for many lifting tasks particularly those performed by novice lifters, in some cases spinal loads can be minimised by using a momentum lifting strategy. In fact Authier et al. (1996) indicated that expert lifters often choose to lift with a ‘jerky’ motion in order to reduce the required effort by using momentum. The momentum strategy enables the initial acceleration of the load during lifting to be performed via the transfer of momentum from the body to the load, thus decreasing the muscular effort required (McGill and Norman, 1985). Initially the lifter extends the back with flexed arms so that the load does not move, the momentum generated by the lumbar extension is then transferred to the load when the arms straighten causing it to be lifted with a ‘jerk’. However, the momentum lifting technique is only appropriate for light loads lifted by experienced and well trained manual material handlers (McGill, 2002).

8) *Utilise rest breaks and/or vary tasks.*

Repetitive or continuous manual handling tasks should be intermittently replaced with rest breaks or tasks that involve different functional requirements. Cumulative loading of the lumbar spine either by repetitive movements or postures held over a long duration are more often related to LBP than acute spinal loading (McGill, 1997; Norman et al. 1998; Callaghan et al., 2001). Even sitting for a long duration has been significantly correlated with LBP (Videman et al., 1990), due to ligamentous tissue creep and the posterior migration of the inter-vertebral disc nucleus. Thus, appropriate rest breaks or varying manual handling tasks enables the structures of the lumbar spine to reduce the spinal injury risks associated with tissue creep and muscle fatigue (McGill, 2002).

9) *Plan and practice the required manual handling tasks.*

Pre planning specific manual handling tasks enables the handler to assess the weight, size and shape of the load and design contingencies for any unexpected perturbations to the load (Sullivan, 1989). Unexpected loading, either by lifting objects of unknown weight or via unexpected perturbations, increase the risk of spinal injury as the body tends to

overcompensate by dramatically increasing muscular activity (van der Burg and van Dieën, 2001) that in turn increases spinal load (Dolan and Adams, 1993). However, the lifting technique inherently adopted during a lifting task depends upon the specific lifting conditions (Burgess-Limerick and Abernethy, 1995, 1998; Burgess-Limerick et al., 2000). Whilst some research (Barker and Atha, 1994) has indicated that practicing manual handling tasks does not produce long term changes in lifting technique, other studies have indicated that regular lifting instruction and practice can reduce spinal load by appropriately altering lifting technique (Scopa, 1993; Schenk et al., 1996; Sedgwick and Gormley, 1998; McGill, 2002).

As there are many mechanistic factors that can influence the risk of injury to the lumbar spine in the workplace, guidelines have been legislated by the appropriate governing bodies in the hope of reducing the risk of injury. As a general rule these guidelines follow the safe manual handling guidelines presented above, which are designed to increase the safety margin between mechanical force and structural tissue limits by either, decreasing the force experienced by the lumbar spine or by limiting the factors that decrease the structural integrity of the spinal tissues.

2.4 OHS guidelines on manual handling

Most legislative assemblies in industrialised nations have formed governing bodies that deal with occupational health and safety (OHS) issues, including manual handling practices. A list of some of the major international OHS bodies is given in Table 2.5. The two international OHS bodies founded by the United Nations (WHO and ILO) often highlight areas of public health that require attention. Both of these bodies have indicated that ergonomic factors in the workplace can cause a significant health risk for workers performing manual handling tasks. However, the relevant national and state bodies of each country implement the legislation intended to reduce the risks associated with manual handling tasks.

Table 2.5: International OHS governing bodies.

OHS Governing Body	Country	Web Address
World Health Organisation (WHO)	United Nations	www.who.int
International Labour Organisation (ILO)	United Nations	www.ilo.org
Occupational Health and Safety Administration (OHSA)	USA	www.osha.gov
National Institute for Occupational Safety and Health (NIOSH)	USA	www.cdc.gov/niosh
Canadian Centre for Occupational Health and Safety (CCOHS)	Canada	www.ccohs.ca
European Agency for Safety and Health at Work (EASHW)	Europe	agency.osha.eu.int
Health and Safety Executive (HSE)	UK	www.hse.gov.uk
Federal Institute for Occupational Safety and Health (FIOSH)	Germany	www.baua.de
National Research and Safety Institute for the Prevention of Occupational Accidents and Diseases (INRS)	France	www.inrs.fr
National Institute of Occupational Health (AMI)	Denmark	www.ami.dk
Swedish Work Environment Authority (SWEA)	Sweden	www.av.se
Finnish Institute of Occupational Health (FIOH)	Finland	www.occuphealth.fi
Occupational Safety and Health Service (OSH)	New Zealand	www.osh.dol.govt.nz

Legislation imposed by national OHS bodies require employers to provide a safe working environment for their employees. In reference to manual handling employers are required to assess the manual handling tasks and to minimise the associated risks. Although OHS guidelines are not always compulsory, Marras (2001) found that companies who adopted comprehensive safe manual handling practices saved 31% to 91% on workers compensation costs. Traditionally, the risks associated with lifting were minimised by imposing weight limits for specific groups of workers based upon gender and age (Davies, 1972). However, this approach does not consider the factors specific to the lifting task such as, the size and shape of the object to be lifted, the individual physical capacity of the worker, environmental space and temperature conditions, and the duration and frequency of lifts. Because there are many factors that can affect the risk of injury during lifting tasks it is difficult to legislate quantitative arbitrary guidelines (such as weight limits) that are appropriate for all lifting tasks.

Consequently, most OHS bodies have implemented guidelines that enable risk assessment to be made on a case by case basis. Although these guidelines have the flexibility that enables tasks to be assessed individually, the qualitative nature of these guidelines introduces a subjective element to the risk assessment of manual handling tasks. In contrast to current qualitative guidelines, research conducted by NIOSH has led to a set of criteria for estimating the risk associated with lifting tasks of a quantitative nature.

The quantitative guidelines recommended by NIOSH are based on risk assessment and intervention strategies intended to reduce compression forces at the lumbar spine and subsequently reduce the risk of spinal injury. In 1981 NIOSH released the ‘1981 NIOSH lifting equation’, which set lumbar spinal compression limits intended to aid in the planning of safe manual handling practices. The equation required, frequency and horizontal and vertical load displacement inputs to predict the maximum weight an individual could carry before the Action Limit (AL) and a Maximum Permissible Limit (MPL) were exceeded. The AL was set at a L4/L5 vertebral compression force of 3433 N, where 75% of women and 95% of men can safely perform the manual handling task. The compression force for the MPL was set at 6376 N, which is a level that only 1% of women and 25% of men can lift safely. NIOSH guidelines recommend that no manual handling tasks be performed where compression forces are above the MPL, whilst tasks involving compression forces between the AL and MPL undergo some form of managerial or ergonomic intervention. Loads below the AL are considered acceptable as they represent only a nominal risk of spinal injury to most workers. In 1991, NIOSH expanded on the 1981 equation to include asymmetrical lifting and hand load coupling considerations (NIOSH, 1991). The 1991 lifting equation predicts a Recommended Weight Limit (RWL) above which lifting tasks are considered too hazardous for the majority of the workforce. Both the 1981 and 1991 NIOSH lifting equations have been based upon psychological, physiological and physical/anatomical research (Snook, 1978; Waters et al. 1993; Hidalgo et al., 1995).

Although there have been concerns expressed about the ability of the NIOSH compression force limits to accurately represent the relative risk of lumbar spinal injury (Hidalgo et al., 1997; Jäger et al., 1999), there is sufficient evidence to suggest that excessive biomechanical force in the lumbar spine is related to lower back pain (Herin et al., 1986; Marras et al., 1993; McGill, 1997; Norman et al., 1998). Although the 1991 NIOSH equation has more potential than the 1981 equation to alert occupational health and safety officers to potentially hazardous manual handling tasks (Waters et al., 1993; Potvin and Bent, 1997), the RWL (like the AL and MPL) is still not sensitive to age, gender and strength considerations. Consequently, other alternative quantitative spinal load limits have been proposed. The lifting limits proposed by Jäger et al. (1991) incorporate both age and gender factors to give spinal compression force limits, below which the risk of spinal injury is minimised. Likewise, Genaidy et al. (1993) recommended limits that are based on strength tables for the various percentages of the population. Genaidy et al. (1993) also proposed that limits should be based on the force that incurs the first incidence of injury, compared to the NIOSH limits that are based upon the force that causes vertebral end plate fracture.

Within Australia, the national governing body overseeing manual handling practices is the National Occupational Health and Safety Commission (NOHSC). Although this body has established the Manual Handling: National Code of Practice (NOHSC, 1990), each state has independent guidelines that govern safe lifting practices in the workforce. A list of the national and state OHS bodies within Australia is given in Table 2.6. Each of these state bodies tend to follow the manual handling recommendations made from the NOHSC, though due to some differences in state laws the implementation of these recommendations may be different.

Table 2.6: Australian OHS governing bodies.

OHS Governing Body	State / Country	Web Address
National Occupational Health and Safety Commission (NOHSC)	Australia	www.nohsc.gov.au
WorkCover NSW (WCNSW)	NSW	www.workcover.nsw.gov.au
Victorian WorkCover Authority (WCV)	Victoria	www.workcover.vic.gov.au
Queensland Government Division of Workplace Health and Safety (WHS)	Queensland	www.whs.qld.gov.au
WorkSafe Western Australia Commission (WSWA)	Western Australia	www.safetyline.wa.gov.au
WorkCover Corporation of South Australia (WCSA)	South Australia	www.workcover.com
Workplace Standard Tasmania (WST)	Tasmania	www.wsa.tas.gov.au
Northern Territory Work Health Authority: (WHNT)	Northern Territory	www.nt.gov.au/dbird/dib/wha
ACT WorkCover: (WCACT)	ACT	www.workcover.act.gov.au

In the 1980s extensive deliberations took place between NOHSC and relevant parties within the workforce, on the development of safe manual handling guidelines. As a result, a Draft National Standard and Draft Code of Practice for Manual Handling was developed and released for public comment in February 1989. The draft was based on the Victorian Department of Labour's Occupational Health and Safety (Manual Handling) Regulations 1988 and the associated Manual Handling Code of Practice. The guidelines which are now part of the National Standard (NOHSC:1001) and National Code of Practice (NOHSC:2005) on manual handling (NOHSC, 1990) had tripartite endorsement from government, industry and trade unions sectors. The principal feature of this national code of practice on manual handling is the multifactorial approach, which involves risk identification, assessment and control measures applied to manual handling tasks. The multifactorial approach was considered to be a more appropriate method than the exclusive use of weight limits alone. Specific employer and employee responsibilities are also set down in the National Code of Practice for Manual Handling.

Initial identification of dangerous manual handling practices is the responsibility of both employer and employee, which comes from analysis of workplace injury records, direct observation by OHS officers and consultation with employees. Once manual handling tasks that place individuals at risk have been identified then a risk assessment of those tasks is undertaken. Assessment of manual handling tasks is based upon numerous parameters including: posture, movement speed and orientation, workstation layout, lift duration and frequency, weight, size and shape of the object, skill and age of the worker, clothing and any other special requirements. Control strategies that may be implemented based on the risk assessment include: worker training, implementation of mechanical manual handling equipment, modification of the object, workplace layout, materials flow, task requirements or the implementation of team lifting strategies.

2.5 OHS guidelines governing the use of team lifting

One strategy recommended by the NOHSC, which can be used to reduce the risk associated with a heavy or awkward lifting task, is to employ the use of a lifting team (i.e. the use of more than one person in the lifting task). However, the guidelines on team lifting provided by the NOHSC are brief at best. The complete guidelines pertaining to team lifting in the Manual Handling Code of Practice state:

“To enable load sharing, lifting partners should be of similar height and build and should be trained in lifting techniques. There should be a person nominated as team leader to coordinate the lift. Team lifting should not be used as a first option in risk control.”

Given that half the lifting tasks required in some workplaces may be performed by teams rather than by individuals (Sharp et al., 1997) the recommendation on team lifting in the National Code of Practice would seem quite limited.

In Queensland the guidelines recommended by the Division of Workplace Health and Safety (WHS) on team lifting are more comprehensive than those put forward by NOHSC. Recommendations by WHS (2000) specify that team lifting should only be used where loads above 16 – 20 kg are to be lifted and mechanical assistance or other control methods are not practical. According to WHS, team lifting should not be used as the first strategy to control the risk of a lifting task, as certain added risks are associated with team lifting that are not present during individual lifting. The added risks associated with team lifting, according to WHS, are due to seven factors:

1. A two-person team can lift only 50 – 80% of the added lifting capacity of the individual team members.
2. Individual coordination of lifting is diminished, due the dynamic lifting requirement to ‘fit in’ with the other team members.
3. The team itself lacks coordination because team members cannot exert force simultaneously.
4. Individuals have a tendency to contribute less effort in team lifting than they do in individual lifting.
5. Two-person teams may attempt to lift more than they are physically capable of, placing themselves at increased risk of musculoskeletal injury.
6. Loads are not shared equally between team members, increasing the risk of injury to individuals lifting a greater percentage of the load.
7. The team is less capable of responding to sudden perturbations to the load during team lifting than a person performing individual lifts.

Due to these seven limitations, WHS recommend that extra preparation above that which is encouraged for individual lifts is undertaken. Specific preparation for team lifting tasks according to WHS should include:

1. Assessing the lift and the number of team members required.
2. Preparing and discussing the intended progression of the lift.
3. Timing the lift, including the use of a countdown.
4. Discussing methods for dealing with unexpected loading.
5. Practice using team lifting aids.
6. Practice of the specific team lifting task.

The added risks associated with team lifting and the special preparation recommended by WHS appears valid. However, in comparison to individual lifting very little empirical research has been conducted on team lifting.

2.6 Team lifting research

Much of the research carried out on team lifting has been conducted from a psychophysical perspective. Psychophysical research conducted by Karwowski (1988), Karwowski and Mital (1986) and Sharp et al. (1995) support the claims by WHS (2000) that a two-person team can lift only 50 – 80% of the added lifting capacity of the individual team members. In contrast, Johnson and Lewis (1989) and Mital and Motorwala (1995) have indicated a tendency for a two-person team to have a greater lifting capacity than the combined sum of the individual lifting capacities of the individual team members. A subsequent psychophysical study by Rice et al. (1995) indicated that the lifting capacity of a team is limited by the weaker of the two-team members. This was further supported by Sharp et al. (1997), who found that the lifting capacity for a team of mixed gender was significantly less than for single gender teams. Consequently, the selection and matching of team members is paramount to the results in this type of experiment. Although results vary from psychophysical studies the majority indicate that team lifting capacity can be detrimentally affected by factors that limit the ability of the team members to work together.

Results from some of the psychophysical research indicates that team lifting would be unlikely to significantly reduce the incidence of spinal injury and LBP, as involving more than one person does not significantly increase lifting capacity (Karwowski and Mital, 1986; Sharp et al., 1995). However, studies by Charney et al. (1991) and Charney (1997) found that the incidence of time off work due to spinal injury was significantly reduced by the introduction of specially trained lifting teams in hospitals. Thus, it appears possible that team lifting can be used as an effective lifting strategy if the team is adequately trained.

Marras et al. (1999) used a biomechanical modelling approach to quantify the spinal loads incurred by the lumbar spine for symmetrical and asymmetrical, individual and team lifting tasks. Results from this study indicated that spinal compression was significantly reduced in the symmetrical team lift when compared to the symmetrical individual lift, but the lateral shear force was significantly greater for the asymmetrical team lift than for the asymmetrical individual lift. Of note in the study by Marras et al. (1999) was that the team members were height matched. Lee and Lee (2001) found that lifting teams with members of unmatched height had significantly lower self estimated lifting capacity than teams with members of matched height. Spinal loads have also been found to be lower in team than individual lifting tasks involving person access covers (Mital and Motorwala, 1995). Varcin-Coad and Barrett (1998) indicated that the spinal load experienced by one member of a two-person lifting team may be dependent on the direction of the hand force applied by the other member of the team, subsequently affecting spinal load. As a result, team lifting per se could not be considered a lifting strategy with greater or lesser risk of injury than individual lifting; rather risk assessment must be performed on a case by case basis.

2.7 Summary

Both the incidence and costs associated with occupational LBP are very high and are related to strenuous manual handling lifting tasks. In fact, spinal injury that occurs as a result of excessive lumbar spinal load during manual handling is the most commonly implicated factor leading to LBP. In order to reduce the risk of spinal injury, guidelines pertaining to the use of manual handling in the workplace are predominantly aimed at reducing the load placed on the lumbar spine.

The implementation of lifting teams in the workplace is routinely used for strenuous or awkward lifting tasks in order to reduce the risk of spinal injury and LBP. Nevertheless, OHS guidelines recommend that team lifting should not be used as the primary method for reducing the risk of injury for strenuous manual handling tasks. However, there is little empirical evidence to support this recommendation. Of the modest amount of research that has been conducted on team lifting most has been performed using a psychophysical framework. Most psychophysical research has found that individual lifting capacity is reduced during team compared to individual lifting tasks. However, Marras et al. (1999) reported lower spinal loads in symmetrical team lifting tasks compared to symmetrical load-matched individual lifts.

Chapter 3

Spinal Model

The purpose of this chapter is to: (i) review and critique the biomechanical approaches that have been used to measure spinal load, (ii) describe the biomechanical spinal model that was developed to estimate elbow, shoulder and L4/L5 loads in the present study and (iii) validate the L4/L5 compression and shear forces estimates from the biomechanical spinal model.

3.1 Biomechanical techniques used to measure spinal load

Spinal load can be determined via direct measurement or indirectly through the use of various biomechanical models. A schematic representation of the relationship between the various biomechanical approaches used to measure spinal load is presented in Figure 3.1. Each approach has advantages and disadvantages that are discussed in the following section.

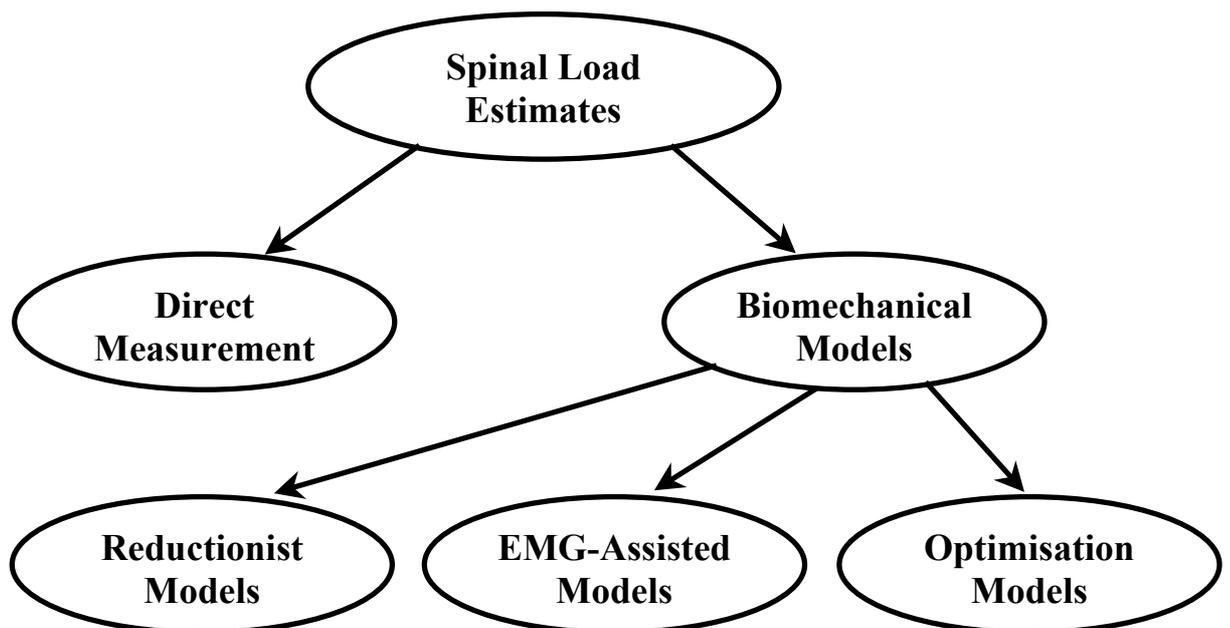


Figure 3.1: Biomechanical methods used to determine spinal load.

3.1.1 Direct measurement

The most valid method for estimating spinal load is via direct measurement. However, the feasibility of using direct techniques is often limited by their invasive nature. In order to directly measure forces in-vivo, the surgical implantation of a transducer is required. Although a limited number of studies have used direct techniques to measure musculoskeletal force in the human triceps surae (Komi, 1990; Gregor et al., 1991; Fukashiro et al., 1993) and the intervertebral compression force of baboons (Ledet et al., 2000), no study has directly measured the forces at the vertebral end plate of a human in-vivo. Direct measurements of intra-vertebral disc and intra-abdominal pressures however, have been obtained (Nachemson, 1981). Although not giving a direct measure of spinal load, intra-vertebral disc pressure does give an indication of the comparative loads experienced by the lumbar spine in different postures (Andersson et al., 1977; Andersson et al., 1982). Likewise, intra-abdominal pressures have been directly measured via an ingested radio pill (Marras and Mirka, 1996). From these studies it has been proposed that intra-abdominal pressure has a load alleviating capacity that reduces the compression force on the lumbar spine by generating an extension torque (Chaffin, 1969; Cyron et al., 1979). More recently, the validity of intra-abdominal pressure as a lumbar spinal load relieving mechanism has been criticised by Gracovetsky et al. (1981) and later by McGill and Norman (1987b). Currently, it is generally accepted that the intra-abdominal pressure during lifting is too low to produce any significant extension torque. Further, to generate large intra-abdominal pressures the abdominal muscles need to be activated, which also produces a flexion torque that is far greater than any extension torque that could be generated by intra-abdominal pressure (McGill and Norman, 1987b; de Looze et al., 1999).

Due to the invasive nature of direct techniques, indirect biomechanical modelling techniques are typically used to estimate spinal load. Using models to study manual handling tasks has the advantage of being able to effectively collect data from multiple

tasks quickly without the experimental injury risks associated with direct methods. Biomechanical models also enable researchers to ask ‘what if ?’ type questions, by varying parameters within the model. Biomechanical models that have been used to estimate the spinal load during various manual handling tasks can be broadly classified as reductionist, EMG-assisted or optimisation models.

3.1.2 Biomechanical Models

Reductionist models

Chaffin (1969) was one of the first researchers to calculate the extensor moment of the lumbar spine during lifting, via a two-link segment model and static Newtonian equations of motion. Chaffin (1975) later extended this model to predict compression forces at the L5/S1 vertebral joint during isometric lifting assuming that a single equivalent muscle force vector produced the lumbar extensor torque. Simple equivalent force vector models have since been used to quantify spinal loads during various manual handling studies (Tsuang, 1992; Mital and Motorwala, 1995; Schipplein et al., 1995; Lavender et al., 2003).

Garg et al. (1982) developed a dynamic model of the lumbar spine by incorporating inertial forces and torques into the static model of Chaffin (1975). Frievalds et al. (1984) suggested that incorporating inertial factors could increase the estimated spinal loads by as much as 40%, compared to static models. McGill and Norman (1985) also compared the estimates from a single equivalent muscle model with and without inertial inputs and found that on average the inertial model predicted lumbar moments 19% greater than the static model.

The first three-dimensional (3-D) analysis of lifting was performed by Schultz and Andersson (1981). Three-dimensional biomechanical spinal models enable the analysis of torques and forces involved with asymmetrical tasks (Dolan and Adams, 1998;

Marras et al., 1999; Skotte et al., 2002). Therefore, 3-D spinal models are important when analysing tasks that involve trunk rotation and lateral bending, which increase the compression load on the lumbar spine due to the activity of the internal and external obliques. However, Bone (1990) found that sagittal plane analysis is often adequate to analyse asymmetrical lifting tasks, as the subject's posture can be out of the sagittal plane by as much as 30 degrees before the difference in lumbar compression force estimates between 2-D and 3-D models exceeds 10%.

Conventional single equivalent force vector models have assumed that the back extensors act parallel to the line of compression through a moment arm of 5 cm (Morris et al., 1961; Chaffin, 1975; Leskinen et al., 1983; McGill and Norman, 1985). These models have often estimated compression forces beyond the vertebral end plate tissue tolerance limits (McGill and Norman, 1987a). More recently, anatomical (Stokes and Gardner-Morse, 1999), medical imaging (Kumar, 1988; McGill, 1988) and moment arm sensitivity studies (van Dieën and de Looze, 1999) have indicated that a single equivalent vector moment arm of 5 cm may underestimate the anatomical and functional capacity of the lumbar extensors to produce torque. As a consequence of these investigations, it is generally accepted that the functional ability of the lumbar extensors is best represented by a 6 cm single equivalent lumbar extensor moment arm (McGill, 1988; Potvin et al., 1991; van Dieën and de Looze, 1999). However, variations in anthropometric characteristics are likely to cause differences in the functional moment arm of the lumbar extensors between individuals.

In addition to the moment arm distance, the line of action of the lumbar extensors has also been investigated. MacIntosh and Bogduk (1987) and McGill and Norman (1987a) both indicated that the line of action of the lumbar muscles and ligaments is dependent on the curvature of the lumbar spine. The results from these studies indicated that when the lumbar spine has a lordotic curvature the lumbar extensors are aligned at approximately a 5-degree posterior angle to the spine and the lumbar ligaments run

parallel to the spinal column. However, when the lumbar spine is fully flexed the lumbar musculature is parallel to the spine and the spinal ligaments have an anterior alignment relative to the spinal column. Due to the effect of the lumbar curvature on the line of action of the muscles and ligaments the shear force is also dependent on the degree of lumbar lordosis in lifting. As a result, lumbar lordosis should be maintained while lifting due to the anterior shear force generated by the interspinous ligaments during full flexion (McGill et al., 2000).

Reductionist models have primarily been used to compare spinal loads between various manual handling tasks as a means of evaluating the relative spinal injury risks associated with each task (Gagnon and Roy, 1992; Gagnon et al., 1992). Although simplistic, errors due to the assumptions made by reductionist models are systematic. Therefore, whilst it may not be appropriate to use spinal load estimates from reductionist models to compare between subjects, reductionist models are valuable tools for comparing the spinal loads that occur for the same subject during different manual handling tasks.

A limitation of reductionist single equivalent muscle models is that they do not consider co-activation of the trunk musculature (Hughes et al., 1994). Single equivalent muscle models will therefore tend to under predict spinal loads when the trunk flexors are activated. However, during symmetrical lifting tasks little activation of the trunk flexors is often present (McGill, 2002). Despite this, the inability of single equivalent muscle models to consider co-contraction has led to the development of models that can account for individual muscle recruitment patterns during manual handling tasks.

EMG-assisted models

To enhance the ability of models to accurately estimate spinal load an approach that combines kinetic, kinematic, and biological (EMG) data has been developed. The development of the EMG-assisted approach has occurred primarily due to the efforts of

Stuart McGill (University of Waterloo) and William Marras (Ohio State University). The development of an EMG-assisted approach to spinal modelling enables more reliable estimates of spinal load to be made during manual handling activities that involve muscle co-activation.

McGill and Norman (1986) produced the first spinal model that incorporated biological (EMG) data. The first part of the EMG-assisted model uses kinematic and kinetic data to give L4/L5 torque estimates based on a link segment model; identical to the first part of the single equivalent muscle model. However, unlike the single equivalent muscle model, McGill and Norman (1986) partitioned the reaction moments between the discs, ligaments and muscular components of the torso. Once the reaction force absorbed by the passive elements are calculated, the remaining force is partitioning amongst 90 torso and lower back fascicles based upon knowledge of the anatomical and functional structure of the muscles and the neural drive to each specific muscle group. The forces generated by the weight of the body and the passive and active elements is then used to determine the compressive and shear force experienced at L4/L5. To calculate the force generated by passive elements kinematic data are used to determine the position of the element on the force-length relationship. Muscular force is calculated via knowledge of: force-length, force-velocity, force per cross sectional area, pennation angle and the activation level of each muscle group as a percentage of maximal EMG via appropriately placed surface electrodes (Winter et al., 1994; McGill et al., 1996). EMG activity is normalised relative to the EMG measured during a maximum voluntary contraction so that the activity of the muscle can be expressed as a percentage of the maximum. A common gain across all muscles is then applied to give the best match to externally measured torque in the lumbar spine. The 3-D adaptation of this model was subsequently used by Pope et al. (1987), McGill and Norman (1987a) and McGill (1992), and the most recent update of the model was given by Cholewicki and McGill (1996).

Granata and Marras (1993) also developed a spinal model that uses EMG data to assist in the prediction of muscle force. This model estimates muscle forces based upon activation and contraction dynamics of the lumbar muscles. Like the model developed by McGill and Norman (1986), this model estimates the neural drive to the muscle as a percentage of the maximal EMG measured during a maximum contraction. Neural drive is then used to directly estimate lumbar torque generated by the lumbar musculature via force-length, force-velocity and cross sectional muscle area data. A gain is then applied to trunk torque estimates so that they match externally measured lumbar torque. Compression and shear forces can then be calculated from the individual muscle force estimates and inertial weight forces.

EMG-assisted models have been extensively used to estimate spinal load during various manual handling tasks (McGill et al., 1995; Marras et al., 1999; Callaghan et al., 2001) and to gain insights into the mechanisms involved in lifting tasks (Granata and Marras, 1995; McGill et al., 1996). The main advantage of the EMG-assisted model is an increased ability to give spinal load estimates that reflect different muscle activation patterns between subjects and tasks demands (Cholewicki et al., 1995). However, with the added individual muscle force information these models offer comes increasing model complexity. The relationship between EMG activity and individual muscle force is not linear during dynamic contractions (De Luca, 1997; Basmajian and De Luca, 1985), making EMG-force predictions difficult. Although non-linear EMG-force relationships have been modelled based on research by Stokes et al. (1987) and Vink et al. (1987), the model by Granata and Marras (1993) assumed the EMG-force relationship to be linear. Furthermore, muscle architecture can also be difficult to model and often varies between subjects. Therefore, the added information that an EMG-assisted model provides, relative to a more simplified model, must be weighed against the increased model complexity and the question that is being investigated.

Optimisation models

An alternative biomechanical modelling approach to quantify spinal loads is via mathematical optimisation. The optimisation approach attempts to predict spinal load based on the assumption that the body is functioning according to a specific objective function that can be minimised. By minimising the objective function the indeterminacy problem can be solved and as a result individual muscle force and consequently the load on the lumbar spine can be estimated (Gagnon et al., 2001). Chaffin (1988), used an optimisation approach that calculated 3-D spinal loads of the lower back whilst minimising intervertebral disc compression and maximising muscle contraction intensity. Although compression force estimates using this objective function are similar to estimates obtained using other objective functions (Hughes, 2000), large discrepancies between spinal load estimates and recorded EMG patterns have been recorded (Chaffin, 1988). Alternatively, Cholewicki and McGill (1994) used an objective function that minimised muscle stress and spinal compression force, whilst calculating the spinal loads in lifting trials. Objective functions that minimise work done (Ayoub, 1998), minimise the sum of cubed muscle stress (Hughes and Chaffin, 1995) and investigate the line of action of the lumbar musculature (Nussbaum et al., 1995) have also been used in optimization models to estimate spinal load. Whilst validating the calculated spinal loads using EMG, Cholewicki and McGill (1994) reported significantly higher spinal load estimates from the optimisation model compared to estimates from an EMG-assisted model. Optimisation techniques have also been used to estimate the force distribution across the lumbar discs (Whyne et al., 1998) and the forces within the lumbar extensors (Calisse et al., 1999) via finite element analysis. Finite element analysis is used to create a 3-D representation of the lumbar spine via the use of structural elements of specific shapes (e.g. triangles, beams and cables) that are fixed to adjacent elements via nodes. Mathematical algorithms can then be used to determine the force experienced at each node, once limiting assumptions that help solve the indeterminacy problem are made.

Whilst the optimisation approach delivers an elegant mathematical solution to the indeterminacy problem, it often violates two important physiological principles. Firstly, these models do not consider individual muscle activation strategies, and thus optimisation models do not explicitly incorporate the effect of muscle co-activation. Secondly, it is unlikely that the minimised objective function used to solve the indeterminacy problem is appropriate for all conditions and all subjects. For example a fatigued subject performing multiple lifting trials is unlikely to be using the same objective function as a power lifter performing a discrete heavy load lifting task. It is also probable that the objective function will change between lifting trials if the intended goal of the individual changes.

Each type of spinal model (reductionist, EMG-assisted or optimisation) has specific advantages and disadvantages. The choice of which model to use therefore depends on the purpose of the investigation being undertaken. Due to the comparative within subject design of the current experiments, a simplistic reductionist model of the lumbar spine was considered the most appropriate model to give spinal load estimates. A detailed description and validation of the spinal model developed for use in the present study is given in the following sections.

3.2 Description of the spinal model

A biomechanical model of the upper body was developed to determine the mechanical loads experienced at the elbow, shoulder and L4/L5 joints during manual handling tasks. Initially, anthropometric, kinematic and kinetic data were used to successively calculate elbow, shoulder and L4/L5 reaction forces and torques. Estimates of L4/L5 reaction force and torque were then subsequently used to determine the compression and shear forces at L4/L5. The spinal model was implemented using customised software (Matlab, version 6.0). A simplified static version of the current spinal model developed using Matlab (version 6.0) is provided in Appendix A.

3.2.1 Estimation of joint reaction forces and net joint torques

A three-linked segment model of the forearm-hand, upper arm and trunk sections was used to represent the mechanical function of upper body during lifting tasks. Each of the three segments was assumed to be rigid and connected by frictionless hinge joints (Figure 3.2). Each segment was also assumed to have a known length, mass, centre of mass location, and rotational inertia about the centre of mass, based upon the regression equations in Winter (1990) (Table 3.1). Whilst errors associated with body segment parameters in Winter (1990) would influence the absolute values of spinal load estimates, they would have no effect on the results obtained in the present study because it utilised a repeated measures within-subject design.

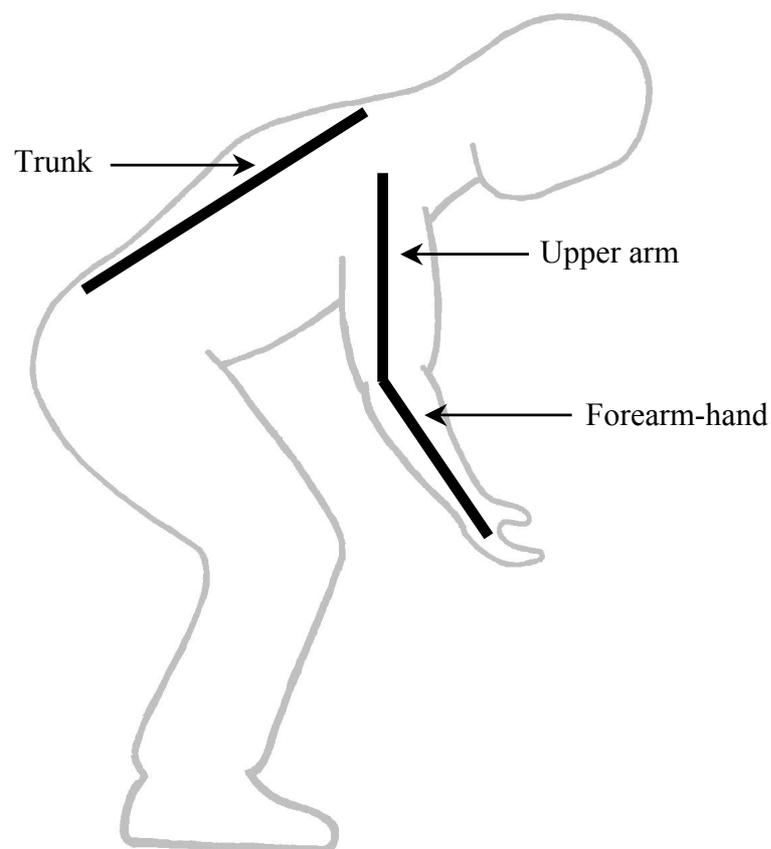


Figure 3.2: Three-link segment model of the forearm-hand, upper arm and trunk.

Table 3.1: Body segment parameter values (adapted from Winter, 1990).

Segment	Definition	SM / BM	CoM / SL	RoG / SL
Forearm-hand	elbow axis – ulnar styloid	0.022	0.318	0.468
Upper arm	glenohumeral axis – elbow axis	0.028	0.564	0.322
Head, neck and trunk	greater trochanter – glenohumeral joint	0.578	0.340	0.503

BM = Total body mass (kg), SM = Segment mass (kg), SL = Segment length (m), CoM = Segment centre of mass from distal end (m), RoG = Segment radius of gyration about the CoM (m)

The moment of inertia about the centre of mass for each segment was calculated via equation 3.1, where I is the moment of inertia, m is the segment mass and r is the radius of gyration.

$$I = \Sigma mr^2 \quad \text{Equation 3.1}$$

The biomechanical model for computing elbow, shoulder and L4/L5 reaction forces and torques was based on the assumption that the task for analysis was sagittally symmetric. That is, the motion of the left and right sides of the body was identical and that all motion was restricted to the sagittal plane. The horizontal and vertical components of the hand load ($F_{hand\ x}$, $F_{hand\ y}$) were assumed to act at the distal end of the forearm-hand segment. Based on a free body diagram of the forearm-hand segment, the elbow joint reaction forces ($F_{elbow\ x}$, $F_{elbow\ y}$) and net elbow joint torque (T_{elbow}) were computed using dynamic equations of motion. This procedure was repeated to compute the shoulder joint reaction forces ($F_{shoulder\ x}$, $F_{shoulder\ y}$) and net shoulder joint torque ($T_{shoulder}$), and finally to compute the L4/L5 joint reaction forces ($F_{L4/L5\ x}$, $F_{L4/L5\ y}$) and net L4/L5 joint torque ($T_{L4/L5}$). Free body diagrams of the forearm-hand, upper arm and trunk are displayed in Figure 3.3.

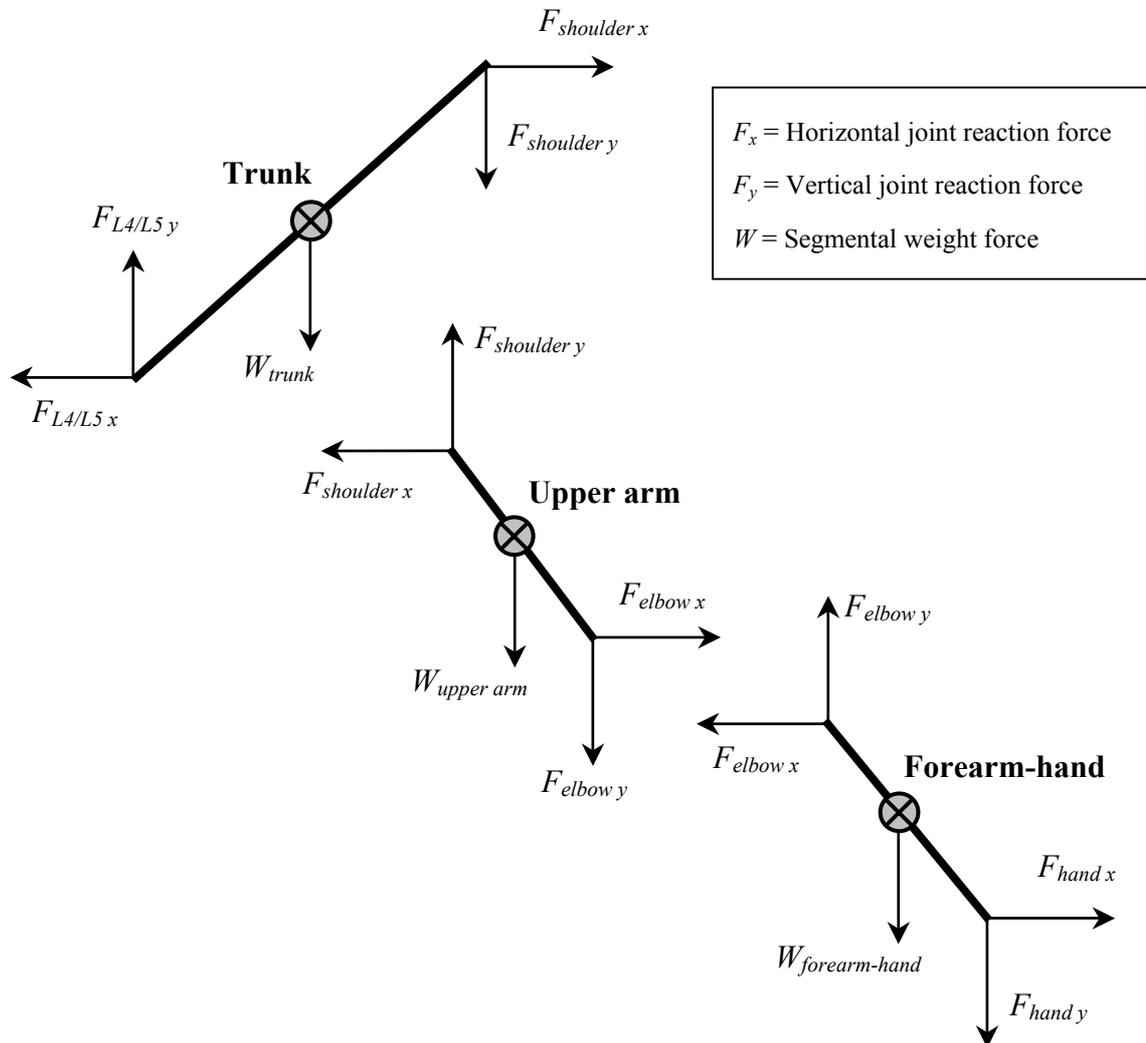


Figure 3.3: Free body diagrams of the forearm-hand, upper arm and trunk segments.

Dynamic Newtonian equilibrium equations were used to estimate the proximal joint reaction forces and net joint moment for each segment (Elftman, 1939). Based on a free body diagram for a generic segment (Figure 3.4), dynamic equilibrium in two dimensions is defined by three equations (Equations 3.2, 3.3 and 3.4) from which any three unknowns may be calculated.

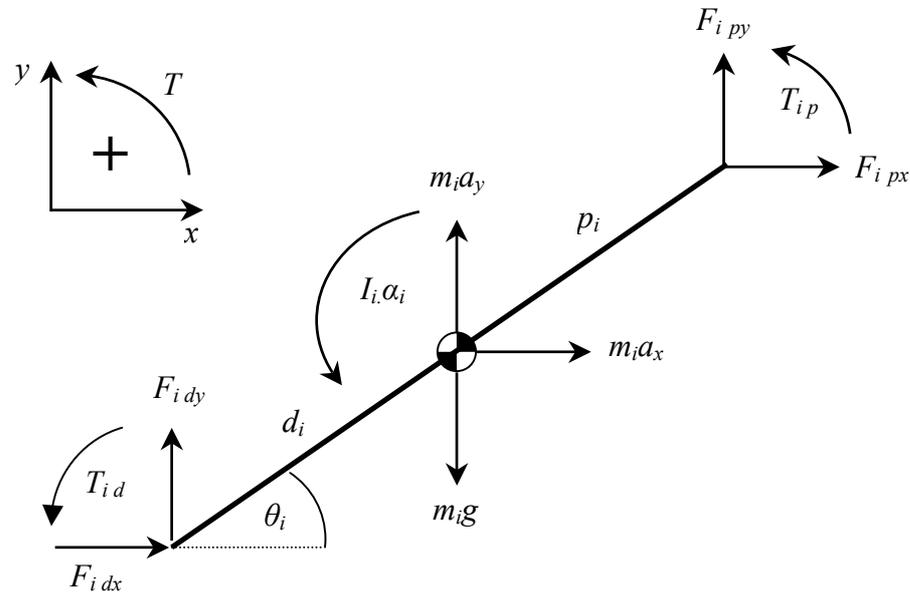


Figure 3.4: Free body diagram for the i^{th} segment.

Nomenclature:

- m_i = mass of the i^{th} segment
- I_i = Moment of inertia of the i^{th} segment about the centre of mass
- p_i = distance between the proximal joint and centre of mass of the i^{th} segment
- d_i = distance between the distal joint and centre of mass of the i^{th} segment
- θ_i = Angle of the i^{th} segment relative to the horizontal
- g = Gravity
- a_x = Linear acceleration of the i^{th} segment in the horizontal direction
- a_y = Linear acceleration of the i^{th} segment in the vertical direction
- α_i = Angular acceleration of the i^{th} segment about the centre of mass
- $F_{i,px}$ = Reaction force in the horizontal direction at the proximal end of the i^{th} segment
- $F_{i,py}$ = Reaction force in the vertical direction at the proximal end of the i^{th} segment
- $F_{i,dx}$ = Reaction force in the horizontal direction at the distal end of the i^{th} segment
- $F_{i,dy}$ = Reaction force in the vertical direction at the distal end of the i^{th} segment
- $T_{i,p}$ = Net torque acting at the proximal end of the i^{th} segment
- $T_{i,d}$ = Net torque acting at the distal end of the i^{th} segment

Using the nomenclature from Figure 3.4, and the Newtonian dynamic equations of motion (Equations 3.2, 3.3 and 3.4), the equations of motion for the i^{th} segment are given below.

Horizontal forces acting on the i^{th} segment

$$\Sigma F_x = ma_x \quad \text{Equation 3.2}$$

$$F_{i_{px}} + F_{i_{dx}} = m_i a_x$$

Vertical forces acting on the i^{th} segment

$$\Sigma F_y = ma_y \quad \text{Equation 3.3}$$

$$F_{i_{py}} + F_{i_{dy}} - m_i g = m_i a_y$$

Torques about the centre of mass of the i^{th} segment

$$\Sigma T_i = I\alpha_i \quad \text{Equation 3.4}$$

$$T_{i_p} + T_{i_d} - (F_{i_{px}} p_i \sin \theta_i) + (F_{i_{py}} p_i \cos \theta_i) + (F_{i_{dx}} d_i \sin \theta_i) - (F_{i_{dy}} d_i \cos \theta_i) = I_i \alpha_i$$

3.2.2 Estimation of L4/L5 compression and shear forces

Once estimates of the torque and reaction forces at L4/L5 were obtained, the next step was to calculate the compression and shear forces experienced at the L4/L5 lumbar spinal joint. Compressive force is defined as the force that is perpendicular to the superior L5 vertebral end plate and shear force is defined as the force that is parallel to the superior L5 vertebral end plate.

L4/L5 compression and shear forces were computed in two steps. First, the single equivalent extensor muscle force (F_M) responsible for the net L4/L5 torque was calculated via Equation 3.5. F_M was calculated assuming that the moment arm of the single equivalent muscle force was 6 cm (McGill, 1988; Potvin et al., 1994; Jorgensen

et al., 2001). Next, the L4/L5 reaction forces were transformed into their compressive and shear components (see Figure 3.5), and the total compressive and shear forces experienced at L4/L5 determined via equations 3.6 and 3.7.

$$F_M = \frac{T_{L4/L5}}{0.06} \quad \text{Equation 3.5}$$

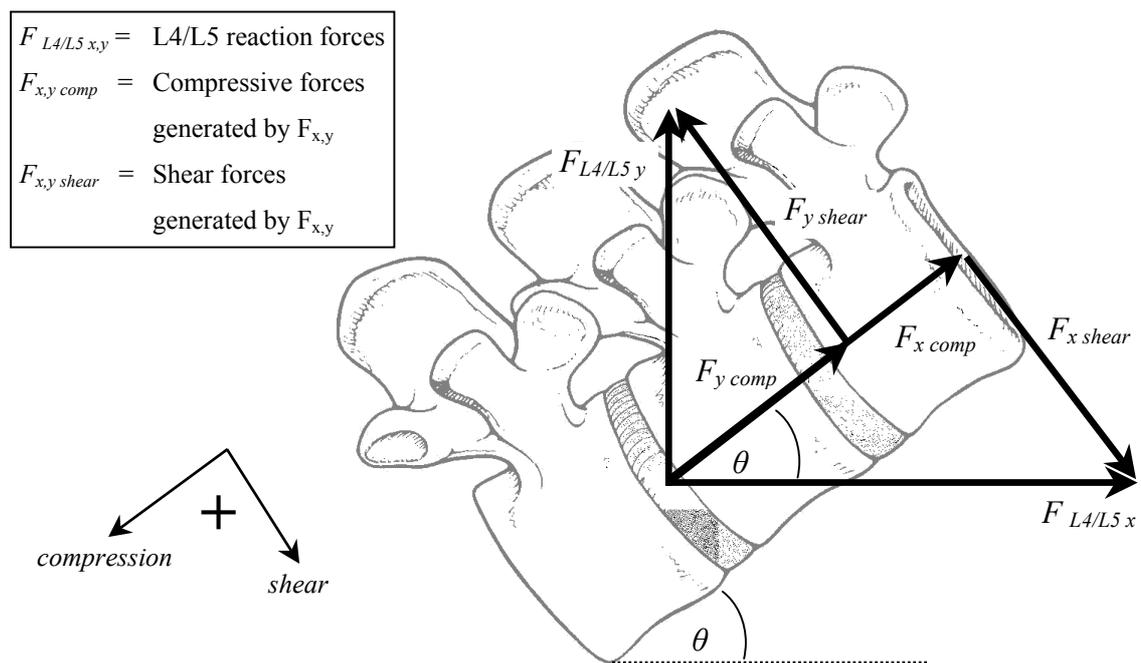


Figure 3.5: Transformation of the horizontal and vertical net L4/L5 joint reaction forces into L4/L5 compression and shear forces.

$$F_{compression} = -(F_{L4/L5x} \cos \theta) - (F_{L4/L5y} \sin \theta) + F_M \quad \text{Equation 3.6}$$

$$F_{shear} = (F_{L4/L5x} \sin \theta) - (F_{L4/L5y} \cos \theta) \quad \text{Equation 3.7}$$

Note: a positive compression force at L4/L5 represents a force that pushes L4 inferiorly relative to L5, whilst a positive shear force represents a force that pushes L4 anteriorly relative to L5.

3.3 Validation of the spinal model

To validate the spinal model, L4/L5 compression and shear force estimates were compared with values obtained using the commercially available ergonomic software programs 3D-SSPP version 2.0 (University of Michigan, 1993) and 4D-WATBAK version 2.0.37 (University of Waterloo, 1999). For a more detailed description of the modelling procedures used by 3D-SSPP and 4D-WATBAK see Stevenson (1998) and Dennis and Barrett (2000). A standard simulated lifting posture, similar to Figure 3.2 (trunk = 60°, upper arm = 0° and forearm-hand = 15° flexion relative to the vertical), of a 1.8 m tall 80 kg male was used as the input into each model. The vertical (150, 200 and 250 N) and horizontal (0, 50 and 100 N) force experienced by the hands was then manipulated and the L4/L5 compression and shear forces were recorded in Table 3.2. Because both 3D-SSPP and 4D-WATBAK are static models, the acceleration inputs into the spinal model presented in this chapter were set to zero to facilitate valid comparisons with the other two models.

Table 3.2: Spinal load estimates from the spinal model, 3D-SSPP and 4D-WATBAK.

VHF (N)	HHF (N)	Compression Force (N)			Shear Force (N)		
		Spinal Model	3D- SSPP	4D- WATBAK	Spinal Model	3D- SSPP	4D- WATBAK
150	0	3342	3325	3368	578	401	494
150	50	3022	2850	3006	603	430	519
150	100	2697	2332	2609	628	459	544
200	0	3797	3695	3778	621	444	537
200	50	3481	3210	3404	646	472	562
200	100	3161	2701	3011	671	502	588
250	0	4253	4062	4187	665	486	581
250	50	3935	3589	3825	689	513	605
250	100	3618	3085	3429	715	544	631

As shown in Table 3.2 the compression and shear force estimates from the spinal model were very similar to the spinal load estimates from both 3D-SSPP and 4D-WATBAK. The average difference (\pm SEM) between the L4/L5 compression force estimates from the current spinal model and estimates from 3D-SSPP and 4D-WATBAK were 273 ± 56 N and 82 ± 20 N respectively. The average difference (\pm SEM) between the L4/L5 shear force estimates from the current spinal model and reaction shear force estimates from 3D-SSPP and 4D-WATBAK were 173 ± 1 N and 83 ± 1 N respectively. Estimates of spinal load by both 3D-SSPP and 4D-WATBAK models were based upon the spinal model developed by McGill and Norman (1986 and 1987a) and the anthropometric data obtained by Plagenhoef (1971), Zatsiorsky and Seluyanov (1983) and Tracy et al. (1989). Therefore, small differences in spinal load estimates between the current spinal model and the two commercially available ergonomic programs may have been due to the simplified nature of the current spinal model and the use of anthropometric equations from Winter (1990). However, the variations in spinal load estimates between the models were considered minor and thus the current spinal model was deemed valid for the intended purpose of comparing spinal loads between the lifting tasks examined as part of the current study.

As is the case for both 3D-SSPP and 4D-WATBAK models, the moment arm of the lumbar musculature in the present model was given a nominal value of 6 cm. This simplified representation of the functional anatomy of the lumbar spine is unlikely to give estimates that are realistic for all subjects. Although this simplified model may not produce estimates that can be used to compare absolute spinal load between subjects, any error is systematic in nature and thus the model can be used to compare spinal loads within subjects under various conditions. Furthermore, the simplified nature of the current spinal model provides a reliability that is harder to obtain with more complex models of the lumbar spine.

Spinal loads during individual and team lifting

Abstract

The aim of this experiment was to compare lumbar spinal loads during individual and team lifting tasks. Ten healthy male subjects performed individual lifts with a box mass of 15, 20 and 25 kg and two-person team lifts with 30, 40 and 50 kg from the floor to standing knuckle height. Boxes instrumented with force transducers measured vertical and horizontal hand forces, whilst sagittal plane segmental kinematics were determined using a video based motion measurement system. Dynamic L4/L5 torques were calculated and used in a single equivalent extensor force model of the lumbar spine to estimate L4/L5 compression and shear forces. A significant reduction in L4/L5 torque and compression force of approximately 20% was found during team compared to individual lifts. Two main reasons for the reduced spinal loads in team compared to individual lifting were identified: (1) the horizontal hand force (i.e. pulling force) was greater in team lifting, and (2) the horizontal position of the hands was closer to the lumbar spine during team lifts. Both the horizontal hand force and position of the hands had approximately equal contributions in reducing the spinal load during team compared to individual lifting.

4.1 Introduction

Team lifting is routinely used as a manual handling strategy in situations where the lifting conditions exceed the safe lifting capacity of an individual and when mechanical assistance is either unavailable or impractical. For example, Sharp et al. (1997) reported

that 53% of all lifts performed by military personnel were performed by more than one person. Similarly, team lifting is commonly used in the healthcare industry to transfer patients (Charney et al., 1991; Daynard et al., 2001) and in the manufacturing and construction sectors (Marras et al., 1999). Although various manual handling guidelines contain recommendations regarding the use of team lifting in the workplace there seems to be limited scientific basis for these recommendations in the literature.

A subjective estimate of lifting capacity known as the maximum acceptable weight limit (MAWL) has been used in psychophysical studies to determine lifting capacity during individual and team lifting tasks (Karwowski and Mital, 1986; Karwowski and Pongpatanasuegsa, 1988; Karwowski, 1988; Johnson and Lewis, 1989; Mital and Motorwala, 1995; Rice et al., 1995; Sharp et al., 1995; Sharp et al., 1997). Though there is some evidence to suggest that the MAWL is related to the risk of low back pain (Snook and Ciriello, 1991), musculoskeletal spinal injury must be caused by excessive mechanical loading to one or more of the spinal structures (McGill, 1997). However, little is known regarding differences in spinal loading between individual and team lifting tasks. It is therefore important to examine and compare spinal loads during individual and team lifting tasks in order to identify factors that may influence the relative risk of injury during team compared to individual lifts.

As well as determining MAWL, Mital and Motorwala (1995) estimated spinal loads during individual and team lifting tasks. Spinal compression forces were found to be higher in team than individual lifting tasks. However, the mass of the load (per person) was heavier in the team than the individual lifting tasks. Therefore, it is difficult to use this investigation to make meaningful comparisons between the spinal loads associated with individual and team lifts.

Several studies have investigated the mechanical loads associated with patient transfers using team lifting strategies. Varcin-Coad and Barrett (1998) estimated spinal loads

associated with three different patient transfer techniques (one individual and two team strategies) and found that one team lifting strategy was associated with the lowest spinal load while the other was associated with the highest spinal load. As a result they concluded that team lifting per se should not be assumed to be associated with either higher or lower spinal load than individual lifts, rather analysis of specific lifting strategies needs to be performed on a case by case basis. Similarly, Lavender et al. (2000) suggested that the optimal patient transfer technique was dependent on the nature of the lifting task.

In the most comprehensive comparison of individual and two-person team lifts to date, Marras et al. (1999) compared 3-D lumbar spinal loads under a range of symmetrical and asymmetrical lifting conditions. Results indicated that lifts performed under asymmetrical conditions incurred significantly higher lateral torque and shear force during team than individual lifting. Conversely, during symmetrical lifting tasks the maximum sagittal plane torque and compression force at L5/S1 were both found to be significantly lower during team than individual lifts. Marras et al. (1999) proposed that the lower sagittal plane L5/S1 torque and compression force during team than individual lifts may be due to ‘less hip tilt and more trunk momentum’, though it was suggested that there may be other factors responsible for the differences in spinal load between individual and team lifting tasks.

For the present study it was hypothesised that one possible factor contributing to the difference between spinal loads during individual and team lifts was the direction of the resultant external force acting on the hand. Whereas the hand force acts primarily in a vertical direction during individual lifts, the hand force may have a significant horizontal component during team lifting. This possibility arises if the two members of the lifting team apply an equal by opposite pulling force on the load. For symmetrical lifting tasks performed with the hands below L4/L5 a horizontal pulling force would generate an extensor torque about L4/L5. The extensor torque generated by the

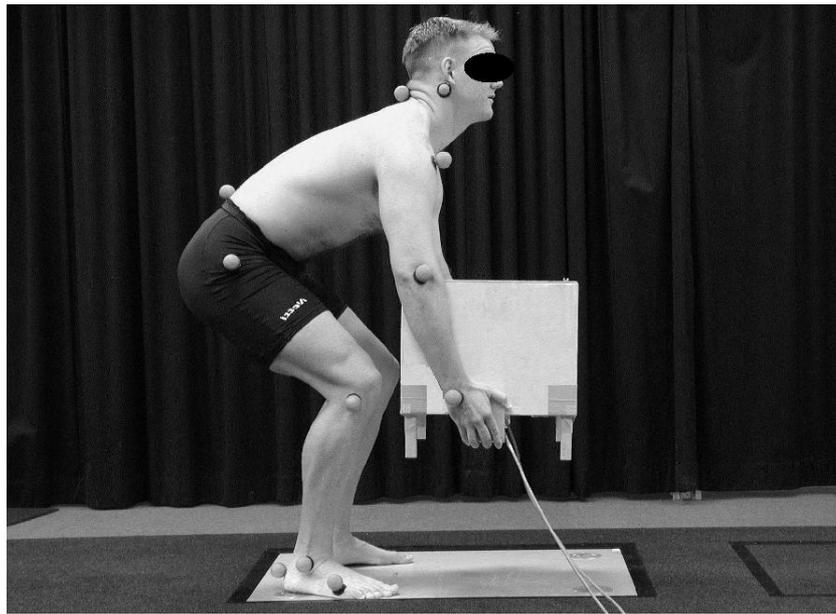
horizontal component of the hand force would reduce the required restorative torque about L4/L5 and therefore also reduce the L4/L5 spinal compression force. The purpose of this study was to compare the sagittal plane L4/L5 torque, compression force and shear force under individual and team symmetrical lifting conditions and to identify the main factors responsible for any observed differences.

4.2 Methods

4.2.1 Subjects and experimental procedures

Ten healthy male subjects (mean \pm SD, age: 25 ± 5 yrs, height: 1.79 ± 0.06 m, weight: 79 ± 9 kg) with no background or specific training in manual handling techniques were recruited from the student population of Griffith University. Each subject performed five individual lifting trials with 15, 20 and 25 kg loads, and five two-person team lifting trials with 30, 40 and 50 kg loads. Subjects in the two-person lifting teams were height and weight matched to within 5%. During the individual lifting trials subjects were instructed to lift a box 50 cm wide, 40 cm long and 30 cm high from the floor to standing knuckle height using their own preferred lifting technique (Figure 4.1a). Team lifts were performed with each member of the two-person team facing each other whilst lifting from the ends of a box that was twice as long (i.e. 50 x 80 x 30 cm) as in the individual lifting trials (Figure 4.1b). Although no specific lifting instructions were given to the subjects in either lifting condition regarding feet position or lifting technique, one member of each two-person team was assigned as team leader who verbally coordinated the timing of the two-person team lift via a '3 - 2 - 1 - lift' countdown. Loads were evenly distributed within the box for both individual and team lifting tasks. At the completion of each lifting trial subjects gave a rating of perceived exertion (RPE) score between 0 and 10 using the Borg CR-10 scale (Borg 1982).

(a)



(b)

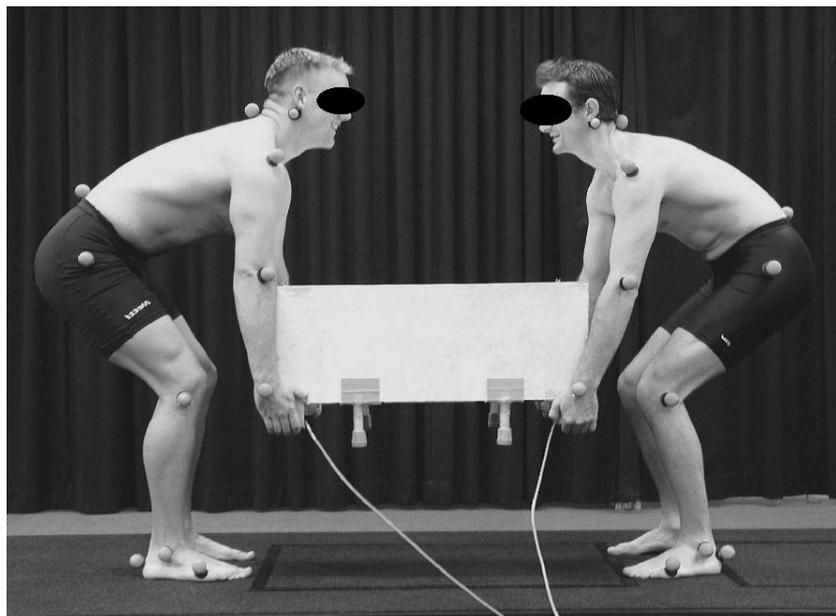


Figure 4.1: Subjects performing an individual (a) and a two-person team (b) lift. The instrumented boxes (which were twice as heavy and twice as long in the team lifting condition) measured horizontal and vertical hand loads via a set of 3-D force transducers that were located in the centre of the box during individual lifts and at the ends of the box during team lifts.

4.2.2 Data collection procedures

Both the individual and team boxes were instrumented with a set of 3-D force transducers (Triaxial load cell, Applied Measurement) which measured dynamic hand force at 1000 Hz. Prior to conducting the experiment it was necessary to determine the normal hand positions used in individual and team lifts so that the force transducers could be mounted on the boxes accordingly. Preferred hand placements were determined from a pilot study involving eight subjects. Each subject performed the same lifting tasks as used in the experiment except that the subjects were able to grip the individual and team boxes in their own preferred manner. During team lifts all subjects placed their hands underneath the adjacent corners of the box, whilst for individual lifts seven of the eight subjects used the centrally symmetrical hand placement shown in Figure 4.1. The one subject who lifted the box using an asymmetrical hand position had previously been trained in manual handling techniques. The advantage of using this asymmetrical technique was that the box was kept closer to the low back than during symmetrical hand placement, although not as close as during the team lifting condition. However, the asymmetrical hand placement technique also involved a substantial amount of non-planar movement, which would have resulted in substantial errors in our 2-D spinal load estimates. Therefore, it was decided to position the force transducers so as to constrain the hand placement used in the individual lift to be the same as the symmetrical position used by seven of the eight subjects in the pilot study. The force transducers were therefore located in the centre of the box during individual lifts and at the ends of the box during team lifts, as shown in Figure 4.1. As a result the hand placement and lifting technique used by the subjects in the present study most likely reflects the individual lifting technique used by persons who have not undertaken manual handling training.

Kinematic data were obtained via reflective markers, which were placed on the head of the third metacarpal, lateral epicondyle of the humerus, acromion process of the

scapula, styloid process of the temporal bone, spinous process of C7 and the spinous process of the L5 vertebrae. Marker trajectories were recorded at 50 Hz and their coordinates obtained using the Peak Motus Motion Measurement System.

4.2.3 Data analysis procedures

Kinematic data were filtered using a dual pass 2nd order Butterworth low pass filter with a cut-off frequency of 6 Hz. Hand forces and kinematic data describing the trajectories of the forearm, upper arm, head and trunk during lifting were used as inputs into customised software (Matlab version 6.0) that calculated the dynamic L4/L5 torque and joint reaction force in the sagittal plane. Compression and shear forces at L4/L5 were calculated assuming a single equivalent extensor force with a moment arm of 6 cm (McGill, 1988; Jorgensen et al., 2001).

In order to identify the causes of any difference in spinal loads found between individual and team lifting the horizontal hand force (HHF), moment arm of the HHF relative to L4/L5 (HHFma), vertical hand force (VHF) and the moment arm of the VHF relative to L4/L5 (VHFma) were calculated (Figure 4.2).

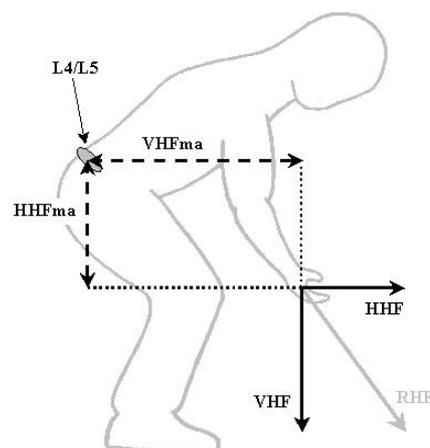


Figure 4.2: Schematic representation of a lifting task showing the horizontal (HHF) and vertical (VHF) components of the resultant hand force (RHF), along with their respective moment arms (HHFma and VHFma) relative to L4/L5.

A general linear model with repeated measures for lift type (i.e. individual and team lifting) was used to determine the effect of the lift type on the dependent measures (RPE, lift duration, maximum and average L4/L5 torque, compression and shear force, and maximum and average HHF, HHFma, VHF and VHFma) averaged across the three loading conditions (i.e. 15, 20 and 25 kg per person). Statistical analysis was conducted using the Statistical Package for Social Sciences (SPSS version 10.0) and significance was accepted at $p < 0.01$. All descriptive statistics are reported as the mean \pm one standard error of the mean (SEM).

4.3 Results

Ensemble averages of the L4/L5 torque, compression force and shear force for the individual and team lifting tasks at the 15, 20 and 25 kg loading conditions (per person) are displayed in Figure 4.3. L4/L5 torque and compression force at all three load conditions remained lower throughout the duration of the team compared to the individual lifting task. Maximum L4/L5 compression forces during team and individual lifts at all three loading conditions were above the NIOSH action limit of 3433 N and below the NIOSH maximal permissible limit of 6376 N (NIOSH 1981). Maximum L4/L5 compression forces can also be compared to the age and gender specific compression force limit proposed Jäger et al. (1991), which for the subjects in this study was 5500 N. When averaged across loading conditions the maximum compression force during team lifting of 4586 N was below the Jäger limit, whilst maximum compression force during individual lifting of 5659 N was above the Jäger limit.

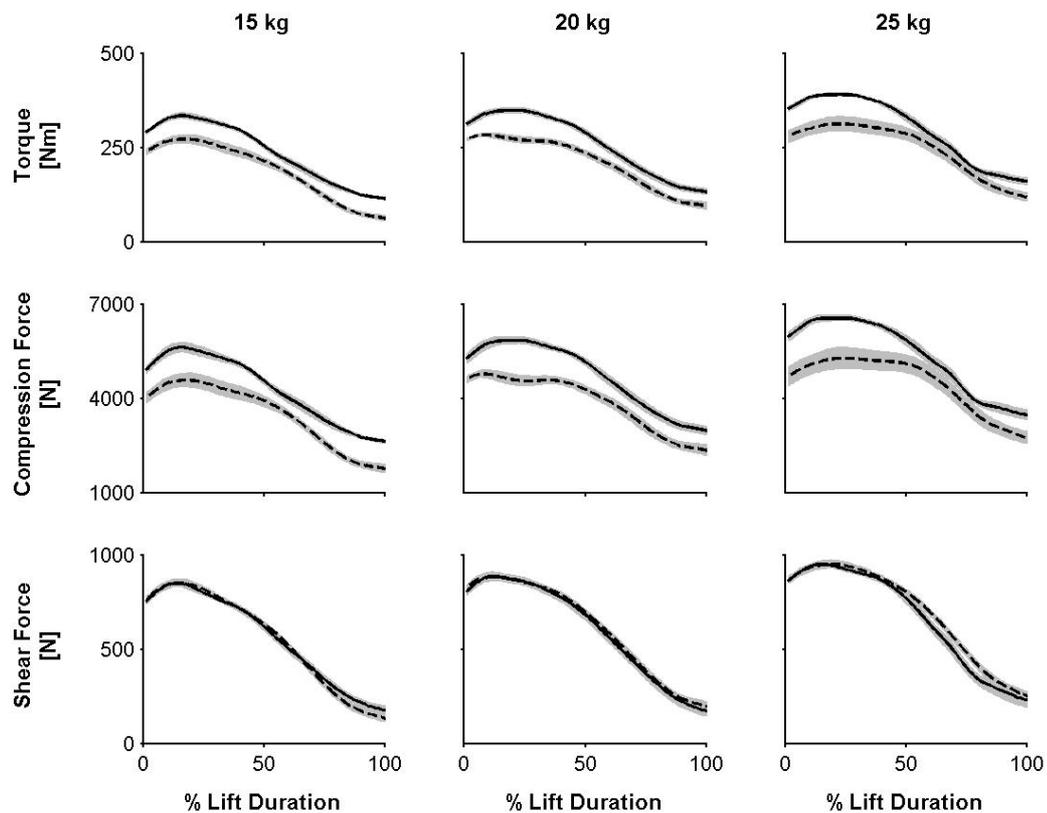


Figure 4.3: Ensemble averages of the torque, compression force and shear force at L4/L5 for the individual (solid lines) and team (dashed lines) lifting tasks at the 15, 20 and 25 kg loading conditions (per person). All data are normalised to lift duration. Shaded area represents the standard error of the mean (SEM).

Statistical comparisons between individual and team lifting spinal loads (maximum and average) averaged across all three load conditions (15, 20 and 25 kg per person) are presented in Table 4.1. Lift duration, maximum and average L4/L5 torque, and maximum and average L4/L5 compression forces were all significantly lower ($p = 0.001$) during team compared to individual lifts. The reduction in the torque and compression force at L4/L5 during team compared to individual lifting ranged between 18 - 20% (Table 4.1). Similarly, comparison of the RPE scores between individual and

team lifting tasks indicated that subjects estimated the effort required to lift the load was 14% lower during team lifting. However, the difference between RPE scores during individual and team lifting was not significantly different ($p = 0.149$).

Table 4.1: RPE, lift duration and spinal load for individual and team lifts averaged across all loading conditions (mean \pm SEM) and their percentage increase or decrease during the team compared to the individual lifting condition.

Variable	Individual Lift	Team Lift	Difference
RPE	3.00 (0.25)	2.58 (0.36)	14% ↓
Lift Duration (s)	1.50 (0.06)	1.33 (0.06)	11% ↓ *
Maximum Torque (Nm)	335 (15)	268 (14)	20% ↓ *
Average Torque (Nm)	240 (10)	194 (11)	19% ↓ *
Maximum Compression (N)	5659 (280)	4586 (254)	19% ↓ *
Average Compression (N)	4340 (175)	3550 (187)	18% ↓ *
Maximum Shear (N)	825 (28)	821 (22)	<1% ↓
Average Shear (N)	543 (17)	565 (16)	4% ↑

* Denotes a significant difference between individual and team lifts ($p < 0.01$).

Ensemble averages of the HHF, HHFma, VHF and VHFma for individual and team lifts at the 15, 20 and 25 kg loading conditions (per person) are displayed in Figure 4.4. The mean HHF at all three load conditions remained higher throughout the duration of the team compared to the individual lifting task. Conversely, the mean VHFma was lower throughout the duration of team compared to individual lifts at each loading condition.

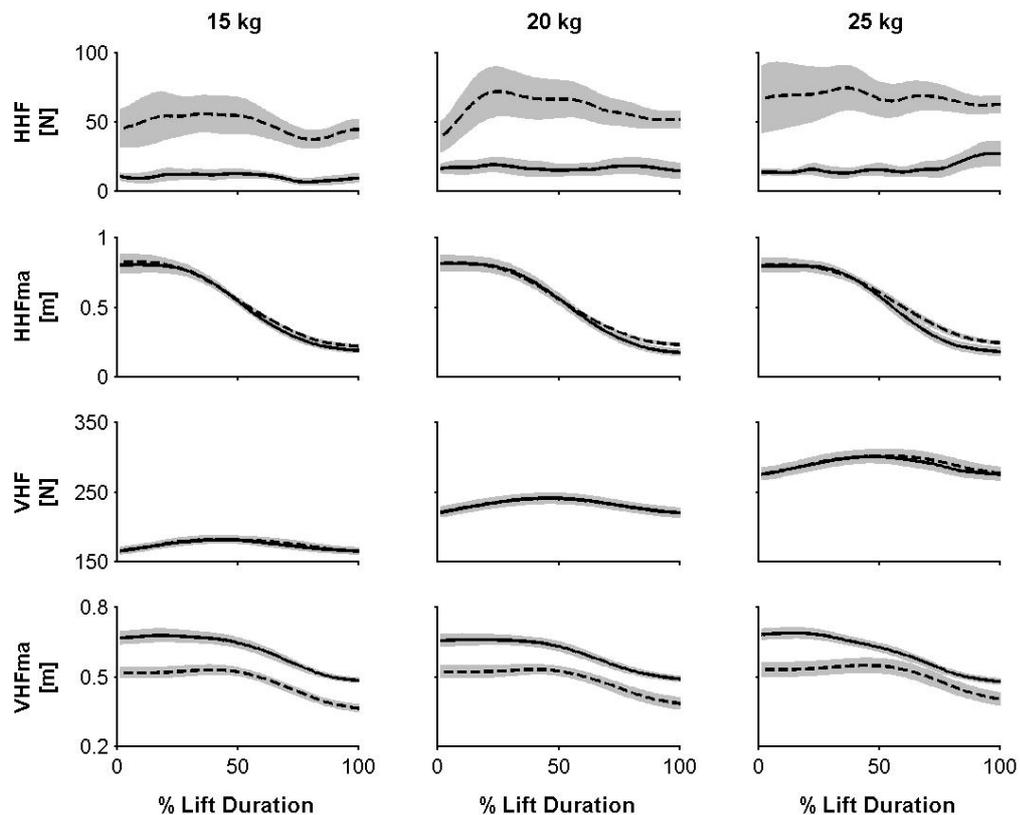


Figure 4.4: Ensemble averages of the horizontal (HHF) and vertical (VHF) hand forces and their respective moment arms (HHFma and VHFma) for individual (solid lines) and team (dashed lines) lifting tasks at the 15, 20 and 25 kg loading conditions (per person). All data are normalised to lift duration. Shaded area represents the standard error of the mean (SEM).

Statistical comparisons between maximum and average HHF, HHFma, VHF, and VHFma in individual and team lifts averaged across all three loading conditions are presented in Table 4.2. A significantly greater maximum ($p = 0.009$) and average ($p = 0.001$) HHF was found during the team compared to during the individual lifting condition. Averaged across all subjects, the maximum HHF was 46 N larger and the average HHF 41 N larger during team than individual lifting trials, which represented a 159% and 315% increase respectively. Conversely, both the maximum and average VHFma were significantly lower ($p = 0.001$) during team than individual lifting. The

lower maximum (0.14 m) and average (0.11 m) VHFma during team compared to individual lifting represented a 22 % and 20 % decrease respectively.

Table 4.2: HHF, HHFma, VHF and VHFma for individual and team lifts averaged across all loading conditions (mean \pm SEM) and their percentage increase or decrease during the team compared to the individual lifting condition.

Variable	Individual Lift	Team Lift	Difference
Maximum HHF (N)	29 (4)	75 (15)	159% \uparrow *
Average HHF (N)	13 (3)	54 (9)	315% \uparrow *
Maximum HHFma (m)	0.72 (0.04)	0.73 (0.04)	1% \uparrow
Average HHFma (m)	0.45 (0.03)	0.49 (0.02)	9% \uparrow
Maximum VHF (N)	217 (1)	217 (1)	<1%
Average VHF (N)	207 (1)	208 (1)	<1% \uparrow
Maximum VHFma (m)	0.63 (0.02)	0.49 (0.02)	22% \downarrow *
Average VHFma (m)	0.55 (0.01)	0.44 (0.01)	20% \downarrow *

* Denotes a significant difference between individual and team lifts ($p < 0.01$).

In order to estimate the relative contributions of the HHF and VHFma to the overall reduction in spinal load during team lifting, the torque generated at L4/L5 by the horizontal and vertical components of the external hand force was calculated by multiplying the average HHF and VHF by their respective moment arms relative to L4/L5. As there was no significant difference in HHFma or VHF between individual and team lifts the reduction in the torque about L4/L5 was primarily due to the HHF and VHFma. The higher average HHF in team lifting acted to reduce the L4/L5 extensor torque by 21 Nm during team compared to during individual lifts. Similarly, the lower average VHFma during team lifts reduced the L4/L5 extensor torque in team compared to individual lifting by 22 Nm. Together the greater HHF and lower VHFma in team lifting reduced the average L4/L5 torque during team lifting by 43 Nm. This 43 Nm

reduction accounts for 96 % of the 46 Nm difference in average L4/L5 torque between individual and team lifting. Therefore, approximately 50 % of the lower L4/L5 torque during team than individual lifts was due to the greater HHF and 50 % due was due to the lower VHFma. Similar results were obtained when estimating the reduction in spinal load during team versus individual lifts on maximum rather than average data.

4.4 Discussion

The present experiment investigated the effect of individual versus team lifting on spinal loads, and sought to identify the mechanical factors responsible for any observed differences.

4.4.1 Spinal loads during individual and team lifting

L4/L5 torque and compression forces were found to be significantly lower in team compared to individual lifts by approximately 20%. This finding is consistent with Marras et al. (1999) who also reported decreased sagittal spinal loads in team lifting relative to individual lifting. The higher absolute spinal loads reported in the present study are probably due to differences in the initial height of the load compared to Marras et al. (1999) who used an initial height of 52 cm.

Maximum L4/L5 compression forces for individual lifts at each loading condition were above the compression force limit proposed by Jäger et al. (1991), which indicates an increased risk of injury to the lumbar spine. In contrast, maximum L4/L5 compression forces during team lifting were below the Jäger limit under all loading conditions and were more than 1000 N lower than the compression forces experienced during individual lifts. Averaged across all load conditions both the individual and team maximum compression forces were above the NIOSH action limit (AL). Loads in excess of the AL are believed to be associated with a moderate increase in risk of injury

to the lower back and warrant administrative or ergonomic intervention (Waters et al., 1993). Results therefore indicate that the individual lift examined in this study is associated with a greater risk of injury to the lower back than the team lift under investigation. However, the finding that spinal loads in team lifting also exceed the AL also reinforce the importance of using mechanical assistance or ergonomic strategies in preference to manual handling.

To assess the influences of inertial loads on predictions of spinal loads during individual and team lifting, linear and angular accelerations of the load and body segments were set to zero. Because the relative differences in spinal loads between individual and team lifts were not affected by the removal of inertial terms it was concluded that differences in lift duration did not influence differences in spinal loads. Furthermore, in accordance with Marras et al. (1999), the reduction in torque and compression force at the lumbar spine during team lifting was not accompanied by an increase in L4/L5 shear force, despite a greater pulling force in team lifting. It can therefore be concluded that sagittal spinal loads are reduced in the team compared to individual lifts assessed in this study.

4.4.2 Factors affecting spinal loads in individual and team lifting

Two main mechanical factors were identified that accounted for the differences in spinal loads between individual and team lifts. These two main factors were the horizontal hand force (HHF) and the moment arm of the vertical hand force (VHFma).

The HHF was significantly greater in the team lifting task than the individual lifting task. If the HHF acting on the lifter is directed away from the body (i.e. if the lifter exerts a pulling force on the load) and the hands are positioned below L4/L5, then the HHF exerts an extension torque about L4/L5. As a result the larger HHF during team lifts reduced the required restorative torque about the lumbar spine and contributed towards a lower L4/L5 compression force in team compared to individual lifting. The

large variability in the HHF during team lifting (Figure 4.4) was primarily due to differences in the HHF between subjects in different teams rather than between subjects in the same lifting team. Whilst the SEM of the maximum and average HHF across all ten subjects was 15 N and 9 N respectively (Table 4.2), the SEM of the HHF between team members was less than half these values at 6 N and 4 N respectively. Therefore, the variability in the HHF displayed in Figure 4.4 represents differences between teams and is unlikely to represent a loss in control or coordination within lifting teams.

Subjects also performed the team lifts with the hands positioned closer to L4/L5 than in the individual lifts. This finding is at least partly explained by the fact that subjects naturally chose to lift from the ends of the box during team lifting and from the centre of the box during the individual lifts as shown in Figure 4.1. The smaller VHF_{ma} during team lifting meant that the vertical component of the external hand force generated a smaller flexion torque about L4/L5 and hence contributed to smaller L4/L5 compression force than during individual lifts.

Approximately equal contributions to the reduction in L4/L5 torque in team lifting were made by the horizontal and vertical components of the external hand force. As neither the HHF_{ma} nor the VHF were significantly different between the individual and team lifting conditions it is concluded that the HHF and VHF_{ma} were the two primary factors responsible for the reduction in spinal load during team lifting.

The subjects that participated in this study had no formal lifting training or lifting experience and were therefore considered novice lifters. Therefore, the hypothesis that subjects intuitively pull the load towards themselves during team lifting was confirmed. The extent to which skilled team lifters may exploit the effects of the HHF and VHF_{ma} during team lifting and the feasibility of training subjects to take advantage of these effects to minimise spinal loads is not known, and thus is an area for future research.

4.5 Conclusion

Maximum and average L4/L5 torque and compression force in the sagittal plane were found to be significantly lower during team than individual lifts. The two main factors responsible for the lower spinal loads during team compared to individual lifting were the higher HHF and lower VHFma in team lifting. Both the horizontal hand force and position of the hands had approximately equal contributions in reducing the spinal load during team compared to individual lifting. Given that subjects in this experiment were not instructed to use a particular lifting technique, it is concluded that subjects naturally exploit the opportunity to decrease spinal loads by exerting a pulling force on the load and keeping the load closer to the lumbar spine in team compared to individual lifting.

Chapter 5

Experiment 2

Spinal loads during two-person team lifting: effect of matched versus unmatched standing height

Abstract

The purpose of this experiment was to compare spinal loads during two-person lifting tasks performed with team members of matched versus unmatched standing height. Twelve young healthy male subjects performed matched and unmatched team lifts with two box masses (30 and 60 kg) and three initial box heights (0, 20 and 40 cm). All lifts were performed in the sagittal plane with a self-selected lifting technique from the initial box height to standing knuckle height. The box was instrumented with force transducers that measured horizontal and vertical hand forces, whilst sagittal plane segmental kinematics were determined using a video-based motion measurement system. Dynamic L4/L5 torques were calculated and used in a single equivalent extensor force model of the lumbar spine to estimate L4/L5 compression and shear forces. No significant differences were found in the maximum torque, compression or shear forces at L4/L5 between lifts performed with team members of matched or unmatched height. However, in the last part of the unmatched lifting condition the box was located closer to the short subject, which caused the reactive L4/L5 torque to decrease for short subjects and increase for tall subjects. Average (but not peak) L4/L5 torque and compression force were therefore significantly lower for short subjects and significantly higher for tall subjects in the unmatched compared to the matched lifting condition. However, the larger spinal load incurred by the tall subjects at the end of the unmatched compared to matched lifts are unlikely to drastically alter the risk of spinal

injury caused by acute loading, because they were only one third of the maximal spinal load that occurred during the lift and were below the NIOSH Action Limit. Despite this matching team member height may still be important for decreasing cumulative loading during repetitive lifting.

5.1 Introduction

Team lifting is a manual handling technique that involves the physical participation of more than one person. Extensive use of team lifting as a manual handling strategy in the workplace occurs because loads are often beyond the maximum lifting capacity of an individual, and either administrative or ergonomic intervention are not applicable. The common use of team lifting in the work place was reported by Sharp et al. (1997), who found that 53% of all lifting tasks performed by military personnel involved the participation of more than one person.

Despite common use, safety guidelines for team lifting are somewhat limited in comparison to the detailed guidelines that have been developed for individual lifting. Throughout the world most occupational health and safety (OHS) regulatory bodies recommend that team lifting should only be used in situations where mechanical assistance is unavailable (NOHSC, 1990; NIOSH, 1994; CCOHS, 1997; WHS, 1999; OSHA, 2000). However, there has been little scientific research conducted to aid in the formulation of safety guidelines specific to team lifting.

Most team lifting studies to date have been conducted from a psychophysical perspective (Karwowski and Mital, 1986; Karwowski, 1988; Karwowski and Pongpatanasuegsa, 1988; Johnson and Lewis, 1989; Mital and Motorwala, 1995; Rice et al., 1995; Sharp et al., 1995; Sharp et al., 1997). These psychophysical lifting studies have typically compared the self-selected maximum lifting capacity of individuals and teams. Although conflicting results have been reported, the majority of these

psychophysical studies have concluded that the maximum lifting capacity of a team is less than the combined maximum lifting capacity of the individual team members.

Though there is some evidence to suggest that self-selected estimates of lifting capacity are related to the risk of injury during lifting (Snook and Ciriello, 1991; Davis and Heaney, 2000), spinal injury itself must be due to excessive biomechanical force beyond the structural limit of biological tissue (McGill, 1997). One method used to assess the relative risk of spinal injury between two lifting tasks is to compare the maximum compression force experienced in the lumbar spine during each lifting task (NIOSH, 1981; Jäger et al., 1991; Waters et al., 1993). Although there have been concerns expressed about the effectiveness of maximum spinal load criteria to accurately represent the relative risk of lumbar spinal injury (Hidalgo et al., 1995; Jäger et al., 1999), there is sufficient evidence to suggest that excessive biomechanical force in the lumbar spine is related to lower back pain (Herin et al., 1986; Marras et al., 1993; McGill, 1997; Norman et al., 1998).

Biomechanical studies by Marras et al. (1999) and Dennis and Barrett (2002) compared lumbar spinal loads during individual and team lifting. Both studies found that during symmetrical lifting tasks the sagittal plane spinal loads were significantly lower during team lifts than individual lifts, even though the mass of the object lifted was twice as heavy in the team than the individual lifting tasks. However, these two studies only examined team lifts where individual team members were of matched standing height.

To our knowledge Lee and Lee (2001) conducted the only study to date examining the effect of relative team member height on the performance of team lifting tasks. Based on subjective evaluations of the maximum load subjects believed they could safely lift, Lee and Lee (2001) reported that the lifting capacity of a two-person team was slightly lower when team members had different standing heights compared to when they had similar standing heights. This significant reduction in the lifting capacity during

unmatched lifting was found even though the average height difference between subjects in unmatched teams was only 4.5 cm. However, this study only examined the total lifting capacity of the team and did not investigate the lifting capacities perceived by the individuals within the lifting team. Furthermore, no study has compared the biomechanical spinal loads incurred during team lifts performed with individuals of matched versus unmatched standing height. Such a study appears worthwhile given that several OHS regulatory bodies recommend that individuals involved in team lifts should be selected so that the team members are of similar standing height (NOHSC, 1990; WHS, 1999; WHS, 2000a). The purpose of this study was therefore to determine the effect of relative team member height (matched versus unmatched) under various object mass and initial height conditions on lumbar spinal loads and the perceived lifting effort required by the individual team members. It was also of interest to identify the causes of any differences in spinal loads found between lifts performed by two-person teams with matched versus unmatched standing height.

5.2 Methods

5.2.1 Subjects and experimental procedures

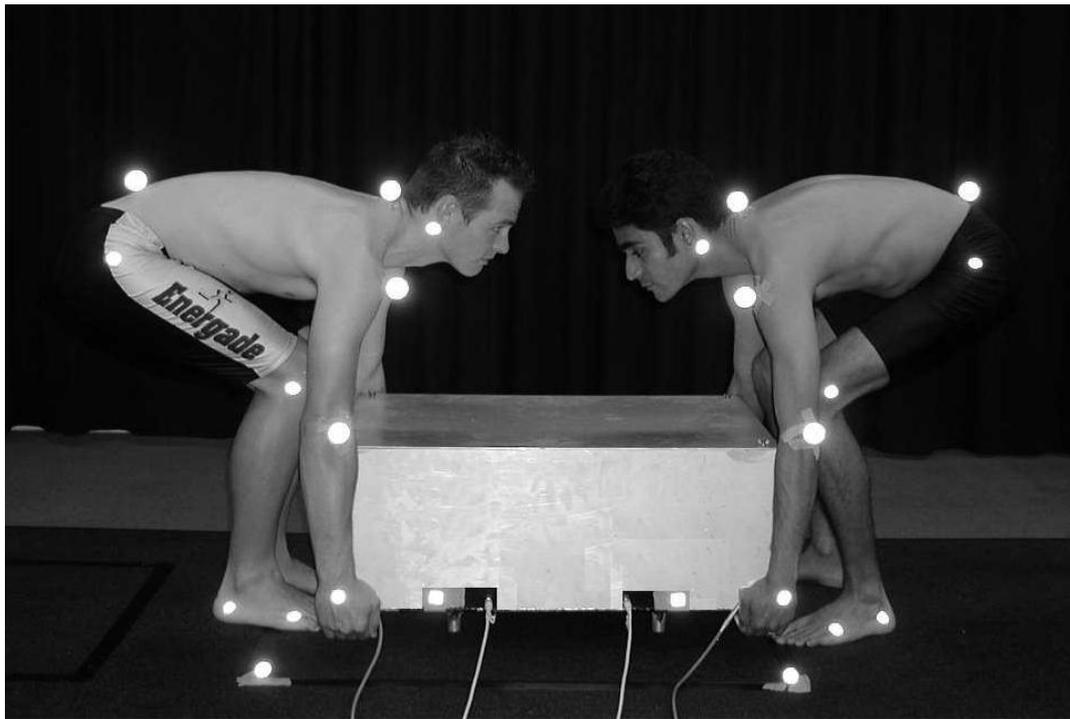
Twelve healthy male subjects with an average age of 25 ± 6 yrs, weight 78 ± 17 kg and standing height of 1.81 ± 0.14 m (mean \pm SD) participated in this study. Six ‘short’ subjects had a standing height less than 1.7 m (26 ± 6 yrs, 63 ± 7 kg, 1.69 ± 0.04 m) and six ‘tall’ subjects had a standing height greater than 1.9 m (25 ± 6 yrs, 92 ± 11 kg, 1.94 ± 0.05 m). The resulting average height difference between the short and tall subjects in the unmatched teams was 25 ± 2.5 cm, and ranged between 23 cm to 29 cm. Subjects were recruited primarily from the student population of Griffith University (Gold Coast Campus), and all experimental protocols were conducted in accordance with ethical clearance granted by the Griffith University Human Ethics Committee.

Team lifts were performed in the sagittal plane with both members of a two-person team facing each other whilst lifting from the ends of a specially built box that was 50 cm wide, 80 cm long and 30 cm high (Figure 5.1). Team lifting tasks were performed with two box masses (30 and 60 kg) and three initial box heights (0, 20 and 40 cm above the floor). Mass was evenly distributed within the box for all lifting conditions. Each subject performed three lifting trials at each condition, both with a team member of matched height (within 5%) and with a team member of unmatched height (Figure 5.1). Consequently, there were 12 different lifting conditions employed in this study. Although no specific lifting instructions were given to the subjects, one member of each two-person team was assigned as a team leader who verbally coordinated the initial timing of the lift via a ‘3 - 2 - 1 - lift’ countdown. At the completion of each lifting trial subjects gave a rating of perceived exertion (RPE) score between 0 and 10 using the Borg CR-10 scale (Borg, 1982), (e.g. 1 = Very weak effort, 3 = Moderate effort, 5 = Strong effort, 10 = Extremely strong effort).

5.2.2 Data collection procedures

The box was instrumented with a set of 3-D force transducers (Triaxial load cell, Applied Measurement), from which dynamic hand forces were sampled at 100 Hz. The transducers were located at the ends of the box in accordance with preferred hand placements identified in pilot testing (Figure 5.1). The initiation of the lifting task by each subject was determined via triggers located at each end of the box. Each trigger was independently activated the instant that end of the box was no longer in contact with the ground. Kinematic data were collected via reflective markers that were placed on the head of the third metacarpal, lateral epicondyle of the humerus, acromion process of the scapula, styloid process of the temporal bone, spinous process of the C7 vertebrae, spinous process of the L5 vertebrae and the corners of the box. Marker trajectories were recorded at 50 Hz and marker coordinates were obtained using the Peak Motus Motion Measurement System.

(a)



(b)

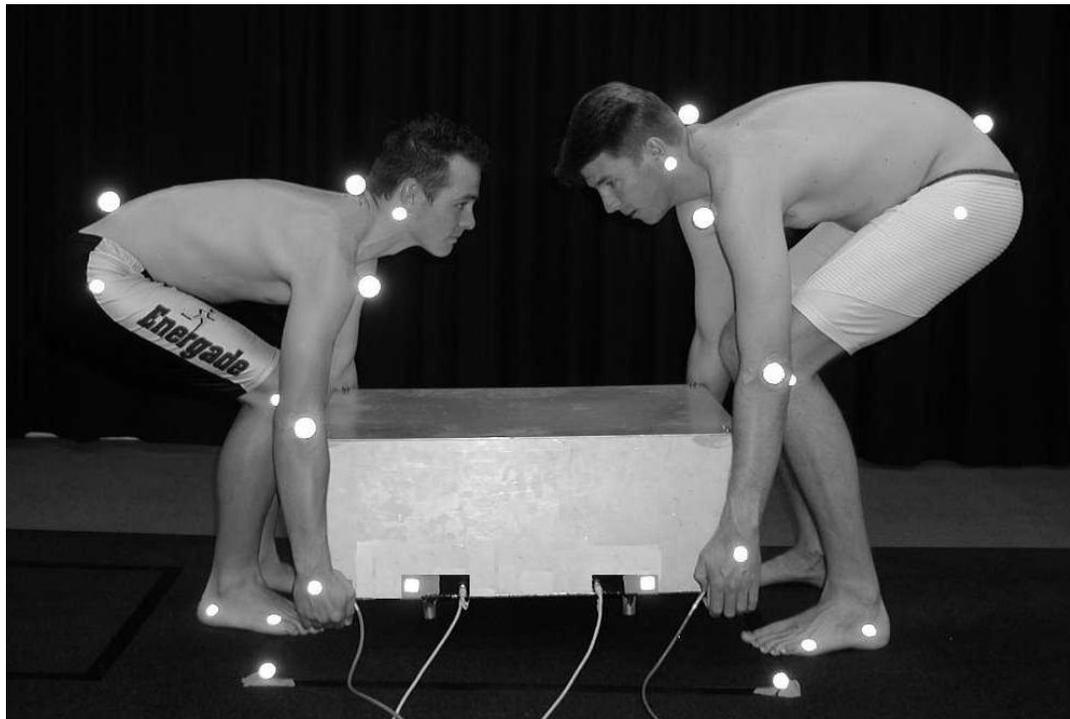


Figure 5.1: A short subject (on the left) performing a matched (a) and unmatched (b) two-person team lift. The instrumented box measured horizontal and vertical hand loads via a set of 3-D force transducers that were located at the ends of the box.

5.2.3 Data analysis procedures

Kinematic data were filtered using a dual pass 2nd order Butterworth low pass filter with a cut-off frequency of 6 Hz. A three-segment sagittal plane model comprising the forearm-hand, upper arm and trunk was used to represent the upper body. Segmental angles were calculated relative to the right hand horizontal and elbow and shoulder joint angles were determined from the angle formed between adjacent segments. Box angle was defined at each end of the box relative to the horizontal (i.e. positive angle indicates an incline and a negative angle indicates a decline of the box from that end of the box to the other end of the box). The temporal coordination between team members at the onset of each two-person team lifting trial was determined by calculating the time difference between the activation of the two triggers located at each end of the box.

Segment lengths, masses, centre of gravity positions and moments of inertia were estimated using regression equations based on subject height and weight (Winter, 1990). Sagittal plane elbow, shoulder and L4/L5 joint reaction forces and torques were calculated using dynamic equilibrium equations based on free body diagrams of the forearm-hand, upper arm and trunk segments. The model was implemented in Matlab (version 6.0) and used hand forces, anthropometric data and kinematic data describing the position and accelerations experienced by each segment as input. L4/L5 compression and shear forces were calculated from the net L4/L5 joint reaction force and torque assuming a single equivalent extensor force with a moment arm of 6 cm (McGill, 1988; Jorgensen et al., 2001).

In order to identify possible factors underlying differences in spinal loads between matched and unmatched lifting, four variables that influenced the restorative torque about L4/L5 were quantified. These variables were the; vertical hand force (VHF), horizontal hand force (HHF), moment arm of the VHF relative to L4/L5 (VHFma) and moment arm of the HHF relative to L4/L5 (HHFma) (Figure 5.2). The sum of the

product of the vertical and horizontal hand forces and their respective moment arms represented the torque generated by the hand load relative to L4/L5.

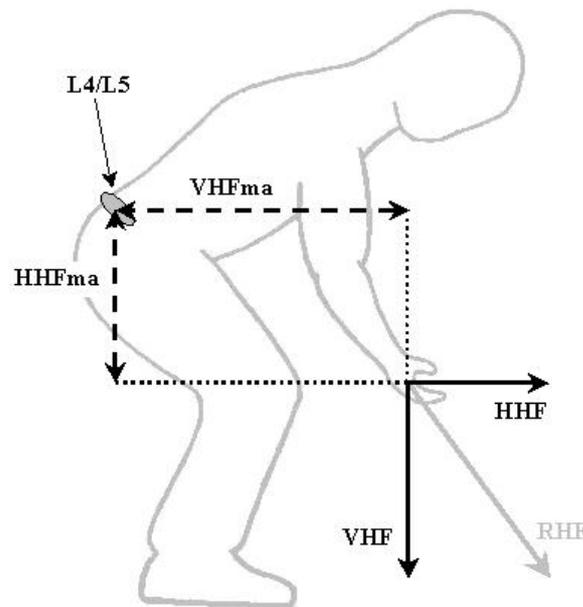


Figure 5.2: Schematic representation of a lifting task showing the vertical (VHF) and horizontal (HHF) components of the resultant hand force along with their respective moment arms (VHFma and HHFma) relative to L4/L5.

5.2.4 Statistical analysis procedures

A mixed $2 \times 2 \times 3 \times 2$ ANOVA with relative team member height (matched and unmatched), box mass (30 and 60 kg) and initial box height (0, 20 and 40 cm) as repeated within-subjects factors and subject height (short and tall) as a between-subjects factor was used to determine significant differences in the dependant measures. Post hoc analysis was conducted using pairwise comparisons with a Bonferroni adjustment ($p < 0.05$). The dependent variables examined were; lift duration, peak and average segment and joint angles, box angle, RPE, peak and average elbow, shoulder and L4/L5 torque, peak and average compression and shear forces at L4/L5, and peak and average VHF, VHFma, HHF and HHFma. The temporal coordination of the lift was examined using a

mixed $2 \times 2 \times 3 \times 2$ ANOVA with the same within-subjects factors, however the team leader variable (i.e. team-leader versus non-team-leader) was used as the between-subjects factor. All statistical analysis was conducted using the Statistical Package for Social Sciences (SPSS version 10.0) and significance was accepted at $p < 0.05$. All descriptive statistics are reported as the mean \pm one standard error of the mean (SEM).

5.3 Results

5.3.1 Joint loading variables

No significant differences in maximum or average elbow or shoulder joint torques due to subject height or team height were detected in the present study. However, differences in joint loading variables were detected for the L4/L5 joint. The spinal loading variables assessed in the present study were maximum and average L4/L5 torque, L4/L5 compression force and L4/L5 shear force. Subject height, box mass and box height had significant main effects on all measures of spinal load (Table 5.1). Tall subjects had significantly higher spinal loads than short subjects. For example, maximum L4/L5 compression forces across all lifting conditions were 51% higher for tall subjects (5809 ± 90 N) than those experienced by the short subjects (3846 ± 70 N). Increasing box mass and decreasing box height also resulted in significantly higher spinal loads. When the box mass was increased from 30 kg to 60 kg, maximum L4/L5 compression force increased by 12% from 4563 ± 137 N to 5092 ± 139 N. Decreasing box height from 40 cm above the ground to ground level (0 cm) resulted in maximum L4/L5 compression force increasing by 6% from 4690 ± 169 N to 4949 ± 168 N. The effects of box mass and initial box height on the maximum spinal loads are consistent with previous research (Waters et al., 1993; de Looze et al., 1996; Granata et al., 1999; Marras et al., 1999).

Table 5.1: Summary of significant F statistics.

Variable	Main effects				Interaction effects	
	subject hgt (short, tall)	rel. team hgt (matched, unmatched)	box mass (30,60 kg)	box hgt (0,20,40 cm)	subject hgt x rel. team hgt	subject hgt x rel. team hgt x box mass
	F(1,10)	F(1,10)	F(1,10)	F(1,10)	F(1,10)	F(1,10)
Joint loading variables						
Max L4/L5 torque	35.45 **		66.24 **	15.60 **		
Avg L4/L5 torque	27.62 **		399.93 **	32.03 **	69.2 **	7.59 *
Max L4/L5 comp.	32.05 **		54.33 **	3.95 *		6.95 *
Avg L4/L5 comp.	27.20 **		411.23 **	10.01 **	41.34 **	
Max L4/L5 shear	22.02 **		280.96 **	74.54 **		
Avg L4/L5 shear	13.31 **		327.46 **	79.30 **		
RPE and lift duration						
RPE	7.83 *		487.51 **	16.19 **	14.18 *	18.73 **
Lift duration				160.23 **		
Factors affecting spinal load						
Max VHF			721.51 **			
Avg VHF			25544.46 **			
Max VHFma	5.84 *			19.25 **		
Avg VHFma	6.23 *			20.85 **	44.02 **	
Max HHF		11.87 **	136.03 **			
Avg HHF		6.71 *	190.30 **		21.01 **	
Max HHFma	16.00 **		8.05 *	109.35 **		
Avg HHFma	23.15 **		25.06 **	76.04 **		
Kinematic variables						
Final trunk angle	9.85 *				17.02 **	
Final box angle	6.21 *	22.45 **			28.62 **	

(* = $p < 0.05$, ** = $p < 0.01$)

While significant main effects were detected for subject height, box mass and box height, no significant main effect due to team height (matched versus unmatched) on any spinal loading variable was observed. However, significant 2-way interactions were detected for the effect of subject height and team height on average L4/L5 torque and average L4/L5 compression force. This 2-way interaction is represented in Figure 5.3, which shows ensemble averages of the L4/L5 torque, compression force and shear force for tall and short subjects in the matched and unmatched lifting conditions. While maximum loads are similar for the matched and unmatched conditions, L4/L5 torque and L4/L5 compression forces are lower for short subjects and higher for tall subjects during the latter part of the lift in the unmatched compared to the matched condition. This results in the average L4/L5 torque and L4/L5 compression force being lower for short subjects and higher for tall subjects in the unmatched compared to the matched condition, although this difference was not evident for average L4/L5 shear forces. Furthermore, a significant 3-way interactions between subject height, team height and box mass were detected (Table 5.1) for both average L4/L5 torque and compression force. This result indicates that the differences in average L4/L5 torque and average L4/L5 compression force for short and tall subjects in the unmatched compared to the matched lifting condition were increased with increasing box mass (Figure 5.4a, b).

The maximum L4/L5 compression force for all subjects under all lifting conditions was above the NIOSH action limit of 3433 N but below the maximum permissible limit of 6376 N (NIOSH, 1981). The highest L4/L5 compression force (6301 N) occurred for tall subjects lifting the 60 kg box from the floor (0 cm) in the unmatched condition. Further, short and tall subjects experienced shear forces above the action limit (500 N) but below the maximum permissible limit (1000 N) proposed by McGill et al. (1998) under all lifting conditions. The highest maximum L4/L5 shear force of 1082 N was experienced by tall subjects lifting the 60 kg box from floor in the unmatched lifting condition, which is just above the maximum permissible limit proposed by McGill et al. (1998).

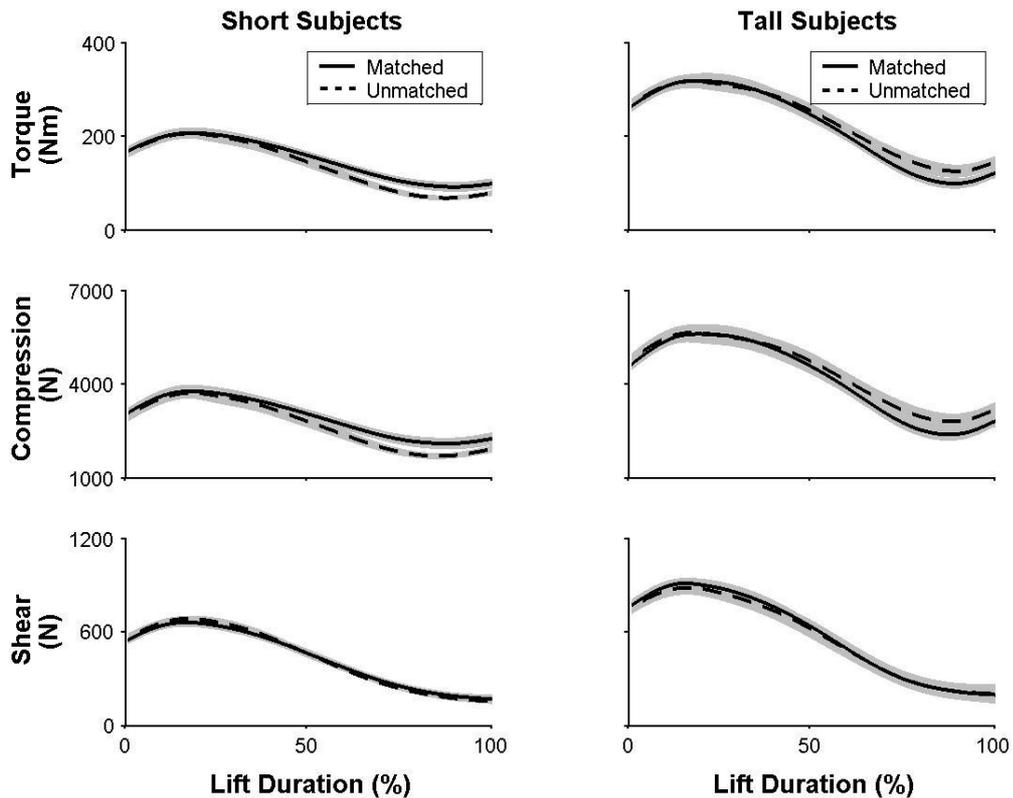


Figure 5.3: Ensemble averages for L4/L5 torque, compression force and shear force for short subjects and tall subjects lifting in matched and unmatched teams averaged across all box mass and initial box height conditions. The shaded area represents the SEM.

5.3.2 RPE

Subject height, box mass and box height all had significant main effects on RPE (Table 5.1). RPE was significantly higher for the tall (2.24 ± 0.18) compared to the short subjects (1.62 ± 0.13), and for the 60 kg condition (3.06 ± 0.11) compared to the 30 kg condition (0.80 ± 0.06). Decreasing box height from 40 cm to ground level (0 cm) caused RPE to increase from 1.57 ± 0.18 to 2.24 ± 0.21 . A significant 2-way interaction between subject height and team height was detected which indicated that RPE was higher for tall relative to short subjects in the unmatched compared to the matched lifting condition. These differences were greater at the heavier box mass condition as indicated by the significant 3-way interaction depicted in Figure 5.4c.

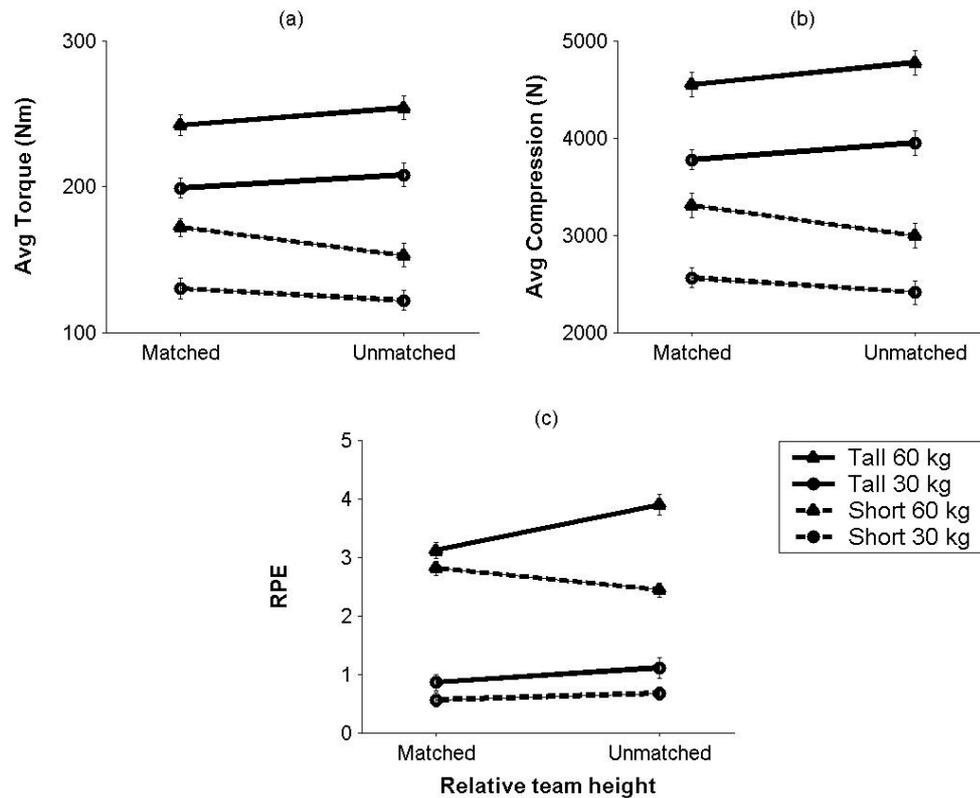


Figure 5.4: Three-way interactions between subject height (short and tall), relative team member height (matched and unmatched) and weight (30 and 60 kg) for the average L4/L5 torque, average L4/L5 compression force and RPE. Error bars represent the SEM.

5.3.3 Lift duration and temporal coordination of the lift

Box height was the only independent variable to significantly affect the duration of the lifting tasks which decreased as the initial height of the box increased from 0 cm (1.46 ± 0.03 s) to 40 cm (1.19 ± 0.04 s). None of the independent variables, including relative team member height, had any affect on the initial temporal coordination of the lift. Collapsed across all lifting conditions there was a 0.06 s average time difference in the initiation of the lift between subjects. Analysis of the lifting initiation times indicated that the team leader commenced the lifts on average 0.02 s before their lifting partner ($F(1,10) = 7.2$, $p = 0.023$). However, no significant difference ($F(1,10) = 0.38$, $p = 0.552$)

was found in the initial coordination of the lift under matched versus unmatched conditions.

5.3.4 Factors influencing spinal loads

A dynamic sagittal plane model of the trunk, upper arm and forearm-hand was used to quantify the joint reaction forces and torques in the elbow, shoulder and L4/L5 joints during selected team lifting tasks. In order to assess the inertial contributions to joint loading, all linear and angular segmental accelerations were set to zero, which resulted in a reduction in the net elbow, shoulder and L4/L5 joint torques by 3%, 3% and 4% respectively. Because inertial forces had a relatively small influence on the magnitude of net joint torques the lifts under investigation were considered to be ‘quasi-static’. The main factors influencing the magnitude of the net L4/L5 torque were therefore the reactive torques produced by trunk, upper arm and forearm-hand segments, as well as the reactive torque produced by the load acting at the hand relative to L4/L5. Because there were no significant differences in segmental and joint angles due to subject height or team height, L4/L5 torques were primarily determined by the torque exerted by the hand load (T_{hand}) about L4/L5 (Equation 5.1). The variables on the right hand side of Equation 1 (see Figure 5.2) were therefore compared across lifting conditions.

$$T_{hand} = VHF \cdot VHFma + HHF \cdot HHFma \quad \text{Equation 5.1}$$

Statistically significant results for factors influencing spinal loading are displayed in Table 5.1. A significant main effect for team height was found for maximum and average HHF, which were 5-10% greater in the unmatched (Max: 90 ± 4 N, Avg: 67 ± 3 N) compared to the matched condition (Max: 82 ± 4 N, Avg: 64 ± 3 N). Main effects were also detected for the effect of subject height and box height on maximum and average VHFma and HHFma, which were greater for tall compared to short subjects and decreased with increasing box height above the floor. Further, maximum and

average VHF and HHF as well as the maximum and average HHFma significantly increased with increasing box mass. Maximum and average HHF almost doubled when the box mass was increased from 30 kg (Max: 60 ± 2 N, Avg: 46 ± 1 N) to 60 kg (Max: 112 ± 3 N, Avg: 85 ± 2 N).

Significant 2-way interactions between subject height and team height for average HHF and average VHFma are displayed in Figure 5.5. Average HHF significantly decreased for short subjects and significantly increased for tall subjects in the unmatched compared to the matched condition (Figure 5.5a). In contrast, average VHFma was significantly increased for short subjects and significantly decreased for tall subjects in the unmatched compared to the matched condition (Figure 5.5b). The average horizontal position of the box relative to L4/L5 was 7 cm closer to the short team member than the tall team member in the unmatched compared to the matched lifting condition. The reason that a significant 2-way interaction for the effect of subject height and team height on maximum HHF and VHFma was not detected is illustrated in Figure 5.6. With the exception of HHF for tall subjects, the ensemble averages for HHF and VHFma are similar for the first half of the lift where maximum values occurred in both matched and unmatched conditions. However, the differences in HHF and VHFma in matched versus unmatched lifting conditions increased throughout the lift.

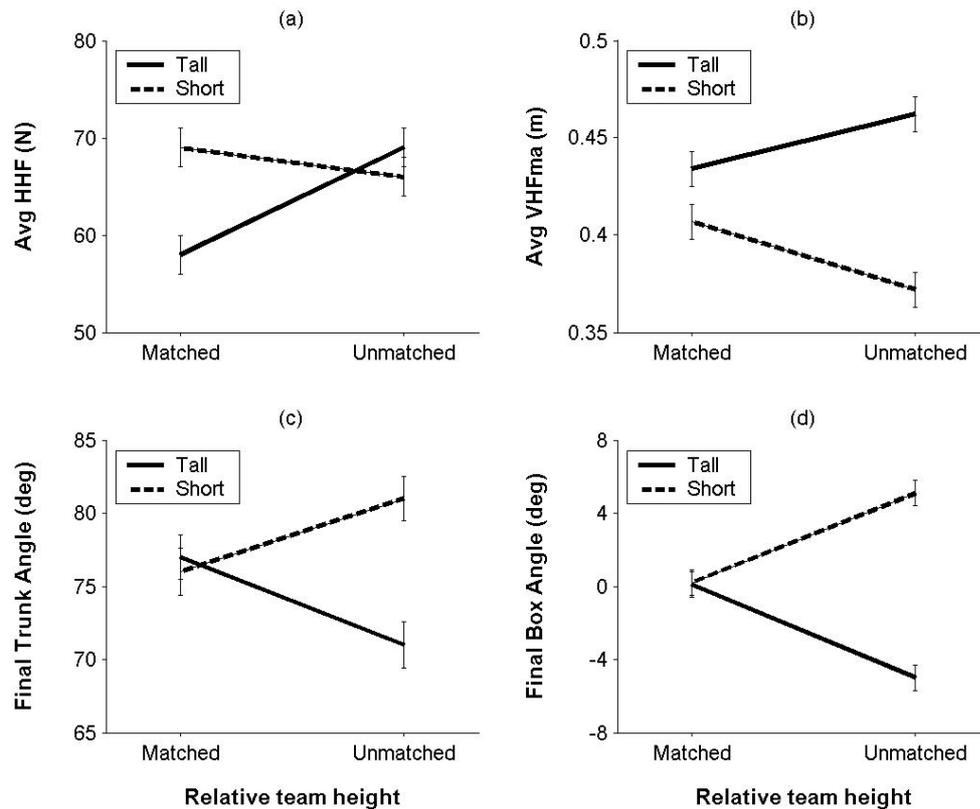


Figure 5.5: Two-way interactions between subject height (short and tall) and relative team member height (matched and unmatched) for the average HHF, average VHFma, final trunk angle and final box angle. Error bars represent the SEM.

5.3.5 Trunk and box angle

Subject height had a significant main effect on trunk and box angles measured at the end of the lift. Trunk angles were significantly higher for short ($79 \pm 2^\circ$) compared to tall subjects ($74 \pm 2^\circ$), indicating that shorter subjects were more upright at the end of the lift. Box angles were significantly higher at the end of the lift for short ($3 \pm 1^\circ$) compared to tall subjects ($-3 \pm 1^\circ$). Team height had a significant main effect on box angle which were greater in the unmatched ($5 \pm 1^\circ$) compared to the matched ($0 \pm 2^\circ$) condition.

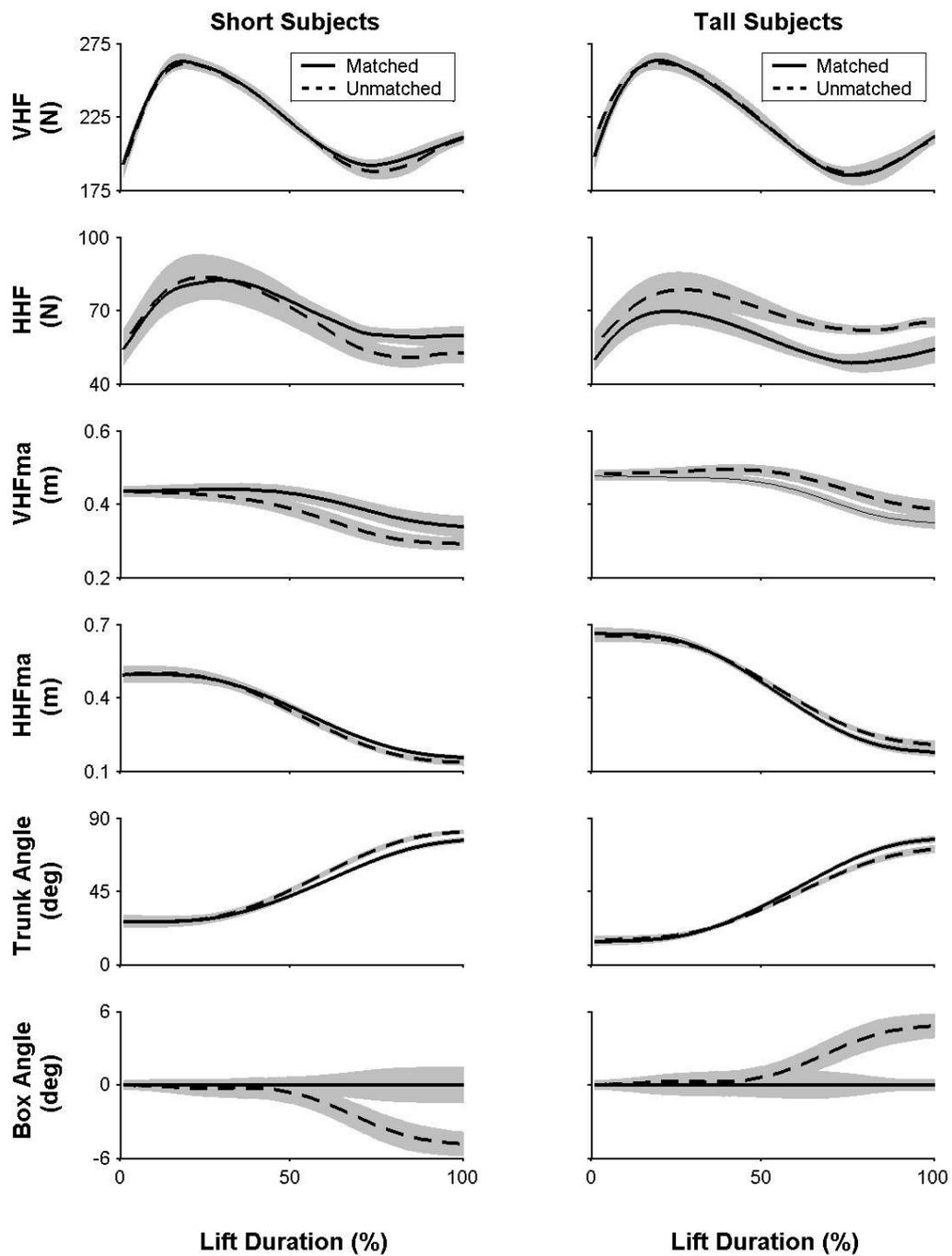


Figure 5.6: Ensemble averages of the vertical and horizontal hand forces (VHF and HHF) and their respective moment arms (VHFma and HHFma), trunk angle and box angle relative to the horizontal for short and tall subjects lifting in matched and unmatched teams averaged across all box mass and initial box height conditions. The shaded area represents the SEM.

Significant 2-way interactions were also detected for the influence of subject height and team height on both trunk angle and box angle (Figure 5.5). The trunk was more upright for short subjects and more flexed for the tall subjects at the end of unmatched compared to matched lifts (Figure 5.5c). Similarly, the box was on average kept level throughout the matched lifts and during the first half of unmatched lifting trials, but tilted (5° maximum) in the latter half of the unmatched lifting condition when the subjects became more upright (Figure 5.5d).

5.4 Discussion

This study sought to identify differences in spinal loads during team lifts performed by subjects of matched and unmatched standing height and to identify the factors underlying any differences found. Because team lifting is a manual handling technique commonly used in the workplace the intention of this study was to help provide scientific evidence to aid the formulation of safety guidelines relating to team lifting practices. Specifically, evidence was sought to evaluate the importance of performing team lifts with team members of a similar standing height.

Contrary to initial expectations, average L4/L5 joint torque and compression force were significantly increased for tall subjects and significantly decreased for short subjects in the unmatched compared to the matched lifting condition. The reason for this difference was that during the latter part of unmatched lifts, the box was positioned closer to the shorter subject, which decreased the reactive torque about L4/L5 for the shorter subject. Associated with this decrease in reactive L4/L5 torque was a concomitant increase in the reactive torque about L4/L5 for the tall subjects in the unmatched compared to matched lifting condition. Given that cumulative loading is a known risk factor for low back injury (McGill, 1997; Norman et al., 1998; Neumann et al., 2001), this result may be important in terms of minimizing cumulative loading of the spine for tall subjects in

regularly performed unmatched lifts. However, differences in spinal loads due to subject and team height were only evident towards the end of the lift where spinal loads were about one-third of maximum loads during the lift and less than the NIOSH Action Limit of 3433 N. It is therefore unlikely that the risk of injury due to acute loading would be markedly affected by lifting in an unmatched team.

Differences in average L4/L5 torque due to the interaction between subject height and team height were related to differences in trunk angle at the end of the lift. Taller subjects remained more flexed than the shorter subjects towards the end of the unmatched compared to matched lifting condition which increased the effective moment arm of the load held in the hands. A likely reason that the tall subjects did not stand fully upright at the end of the unmatched lifting trials was in an attempt to keep the box relatively level. At the end of the unmatched lifting tasks the average height difference between each end of the box was 6 cm, whilst the average difference in standing height between short and tall subjects was 25 cm. If the tall subjects had performed the lift to a more upright trunk position it is probable that this 6 cm height difference would have been much larger.

Consistent with findings for average spinal loads, taller subjects reported significantly higher ratings of perceived exertion than short subjects in the unmatched compared to the matched lifting condition. It therefore appears that for the lifting tasks performed in this study differences in RPE due to the combined effect of subject and team height were related to the changes in spinal loads at the end of the lift.

Although subject and team height significantly affected average L4/L5 torque and compression force, no such differences were detected for maximum L4/L5 torque and compression force. Results therefore indicate that matching the standing height of team members is not an important consideration in terms of minimising the maximum spinal loads associated for the sagittal plane team lifts examined in this study. This would

indicate that the likelihood of spinal injury from an acute load is no greater for the unmatched compared to the matched lifts under investigation.

5.5 Conclusions

Results of the present study indicate that subject height (tall versus short) and team height (unmatched versus matched) interact to influence spinal loads in team lifting. The higher spinal loads for tall compared to short subjects in unmatched compared to matched lifts occurred during the latter part of the lift where spinal loads were well below the peak values attained earlier in the lift. Differences in risk of spinal injury from acute loading between subject and team height conditions are therefore likely to be minimal. If however the team lifts were performed repetitively, the higher average spinal loads experienced by tall compared to short subjects in unmatched compared to matched lifts are likely to increase cumulative spinal loading with possible implications for injury for the taller team member. However, caution is warranted in generalising the results of the present study to other team lifting tasks because other factors such as asymmetrical lifting, space limitations, shape and size of the object to be lifted and the strength, experience and coordination of team members may affect the ability of unmatched teams to safely perform lifting tasks. Further research will therefore be required before definitive conclusions about the importance performing team lifts with team members of a similar standing height can be made.

**Spinal loads during two-person team lifting:
effect of load mass distribution**

Abstract

The purpose of this study was to examine the relationship between load mass distribution (LMD) and spinal load during team lifting tasks. Two-person lifting teams were required to lift a box containing a mass of 30 kg or 60 kg from the floor to standing knuckle height. Adjusting the position of the centre of mass within the box by ± 15 cm and ± 7.5 cm relative to the evenly distributed position (0 cm) yielded three LMD ratios (69:31, 59:41 and 50:50), which represented the percentage of the total mass lifted by each team member. The external force acting on the hands and sagittal plane segmental kinematics were measured and used in a simple dynamic biomechanical model to calculate the torque, compression and shear forces experienced at L4/L5. Spinal load estimates (i.e. maximum and average L4/L5 torque, compression force and shear force) significantly increased with load mass and were positively correlated with LMD ($r = 0.86 - 0.99$, $p < 0.05$), indicating that the person at the heavier end of the asymmetrical load experienced higher spinal loads than their lifting partner. However, the asymmetry in spinal loads between the two team members was found to be significantly lower than the asymmetry in the LMD ratios. This result was primarily due to: (i) a significant positive correlation between LMD and the horizontal “pulling” force acting on the hand and (ii) a significant negative correlation between LMD and the moment arm of the vertical force acting through the hand relative to L4/L5. Thus, when lifting an unevenly balanced load a two-person lifting team seems to adopt a lifting strategy that partially alleviates the larger spinal loads experienced by the team member at the heavier end of the load.

6.1 Introduction

Guidelines on safe lifting practices recommend that the stability of a load be considered when assessing the risks associated with a manual handling task (NOHSC, 1990; WHS, 1999; HSE, 2000). This stability can be compromised when lifting bulky or unwieldy objects or when the contents of the object are likely to shift (HSE, 2000; WHS, 2000b; Lee and Lee, 2002). Further, if one side of the object is heavier than the other the unevenly distributed load may limit the ability of the lifter to react to perturbations of the load or may require uneven or excessive muscle activity to maintain load stability (OSH, 2001). Consequently, uneven distribution of load mass has been identified as a factor that may increase the risk of spinal injury during lifting (CCOHS, 1997; WHS, 2000c; OSH, 2001).

Despite the general preference for manual handling tasks to be performed with evenly balanced loads, there are many situations that require objects with an uneven mass distribution to be lifted (e.g. furniture, machinery). The possible instability associated with moving such objects individually may be overcome via the use of a lifting team. A lifting team has a larger base of support than individuals and thus the instability of the load may be restrained even though the weight of the object may still be unevenly distributed between team members. In fact the percentage of the load that each team member handles may not be equal even when the mass of the load is evenly distributed within the lifted object (Davies, 1972; WHS, 2000a).

Although previous studies have examined a number of factors associated with team lifting including; psychophysical maximal lifting capacity (Karwowski and Mital, 1986; Karwowski, 1988; Karwowski and Pongpatanasuegsa, 1988; Johnson and Lewis, 1989; Mital and Motorwala, 1995; Rice et al., 1995; Sharp et al., 1995; Sharp et al., 1997), spinal loads in height matched teams (Marras et al., 1999; Dennis and Barrett, 2002) and the effects of lifting with team members of unmatched height (Lee and Lee, 2001;

Dennis and Barrett, 2003), no study has investigated the effects of load mass distribution on team lifting. It is likely that spinal loads experienced by individuals in lifting teams are strongly correlated with load mass distribution (LMD). However, it is possible that the coordination strategy adopted by the lifting team might change with LMD, altering the spinal loads experienced by the individual team members. Therefore, the purpose of this study was to examine the relationship between load mass distribution and the spinal loads incurred by the individual team members during two-person team lifting tasks.

6.2 Methods

6.2.1 Subjects and experimental procedures

Twelve healthy male subjects with an average age of 25 ± 6 yrs, weight 78 ± 17 kg and standing height of 1.81 ± 0.14 m (mean \pm SD) were grouped into six two-person lifting teams. Each team was selected so that there was less than 5% difference in the standing height of team members. Subjects had no formal training in manual handling and were recruited primarily from the student population of Griffith University. Experimental protocols were conducted in accordance with the ethical clearance granted by the Griffith University Human Ethics Committee.

Each two-person team performed all lifts in the sagittal plane with a purpose built box (50 cm wide, 80 cm long and 30 cm high) that enabled the distribution of mass within the box to be manipulated (Figure 6.1). Load mass distribution (LMD) was scaled by adjusting the location of the mass within the box so that the centre of mass of the load was positioned ± 15 cm and ± 7.5 cm from the evenly distributed position (0 cm). This resulted in theoretical LMD ratios of 68.7% to 31.3% at ± 15 cm, 59.3% to 40.7% at ± 7.5 cm and 50% to 50% at 0 cm. These LMD ratios are hereafter rounded to 69:31, 59:41 and 50:50. Three lifting trials were performed with two box masses (30 and 60

kg) at each of the five LMD (+15, +7.5, 0, -7.5 and -15 cm) resulting in a total of ten different loading conditions. Therefore, each subject experienced two lifting conditions at the lighter end of the load and two at the heavier end of the load as well as lifting when the load was evenly distributed within the box. No specific lifting instructions were given to the subjects regarding lifting technique. However, one member of each two-person team was assigned as the team leader who verbally coordinated the initial timing of the lift via a ‘3 - 2 - 1 – lift’ countdown. At the completion of each lifting trial subjects gave a rating of perceived exertion (RPE) score between 0 and 10 using the Borg CR-10 scale (Borg, 1982), (e.g. 1 = Very weak effort, 3 = Moderate effort, 5 = Strong effort, 10 = Extremely strong effort).

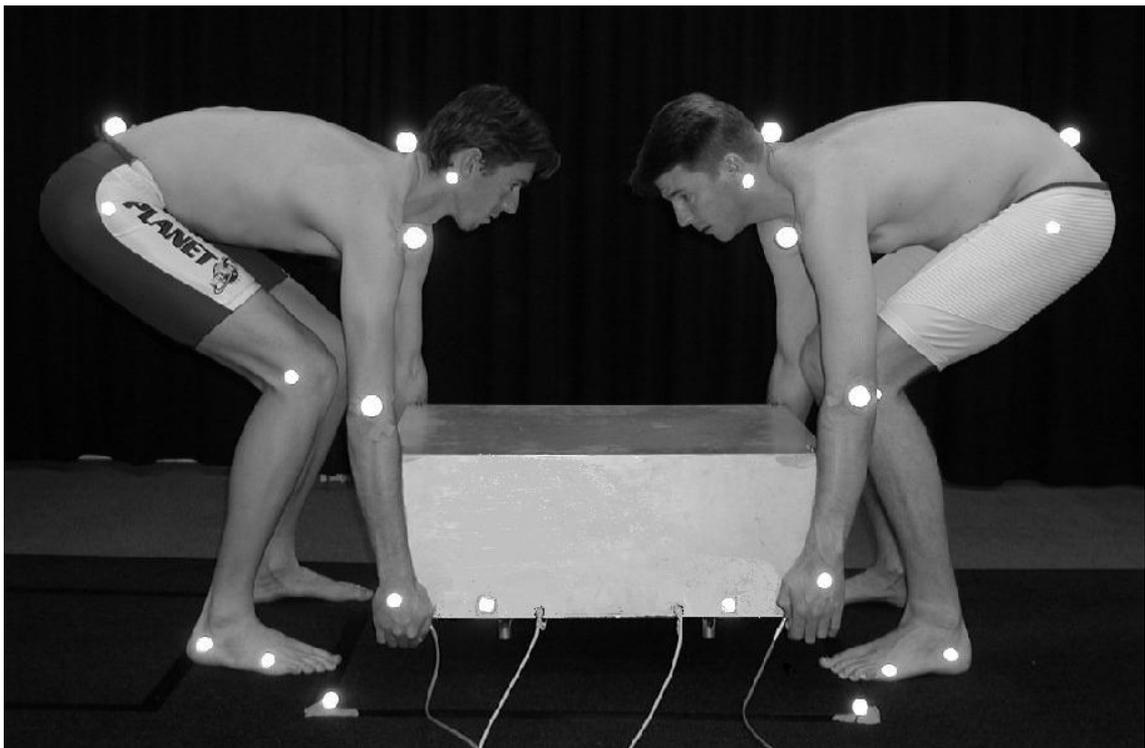


Figure 6.1: Two-person team lift with a 50 cm wide, 80 cm long and 30 cm high box. Three-dimensional force transducers located at the ends of the box were used to measure the forces acting on the hands of each subject.

6.2.2 Data collection procedures

The box was instrumented with a set of 3-D force transducers (Triaxial load cell, Applied Measurement), which were used to measure the vertical and horizontal components of the force acting on each hand at 100 Hz. These forces are subsequently referred to as the vertical hand force (VHF) and the horizontal hand force (HHF). The force transducers were located at the ends of the box in accordance with preferred hand placements identified in pilot testing (Figure 6.1). The initiation of the lifting task by each subject was determined via triggers located at each end of the box. Each trigger was independently activated the instant that end of the box lost contact with the ground. Kinematic data were collected via reflective markers placed on the head of the third metacarpal, lateral epicondyle of the humerus, acromion process of the scapula, styloid process of the temporal bone, spinous process of the C7 vertebrae and the spinous process of the L5 vertebrae. Marker trajectories were recorded at 50 Hz and their coordinates obtained via the Peak Motus Motion Measurement System.

6.2.3 Data analysis procedures

The initial temporal coordination between team members at the onset of each two-person team lifting trial was determined by calculating the time difference between the activation of the two triggers located at each end of the box. Kinematic data were filtered using a dual pass 2nd order Butterworth low pass filter with a cut-off frequency of 6 Hz. Hand forces and kinematic data describing the trajectories of the forearm, upper arm, head and trunk during each lifting trial were used as inputs into customised software (Matlab version 6.0) that calculated the dynamic L4/L5 torque and joint reaction force in the sagittal plane at 100 Hz. Dynamic compression and shear forces at L4/L5 were calculated assuming a single equivalent extensor force with a moment arm of 6 cm (McGill, 1988; Jorgensen et al., 2001). Maximum and average hand forces (VHF and HHF) and spinal loads (L4/L5 torque, compression force and shear force) for

each team member at each lifting condition were expressed in absolute terms and as a percentage of the team's total hand forces and spinal loads. The moment arms of the VHF and HHF relative to L4/L5 (VHFma and HHFma) were also determined.

6.2.4 Statistical analysis procedures

Pearson product moment correlation coefficients were used to examine the relationship between load mass distribution (-15, -7.5, 0, +7.5 and +15 cm) at two box masses (30 and 60 kg) and the dependent measures (maximum and average L4/L5 torque, compression and shear force, RPE, initial temporal coordination, and maximum and average VHF, HHF, VHFma and HHFma). In order to assess the relationship between hand load asymmetry and spinal load asymmetry, the slopes of the regression lines for the percentage of total team hand load and the percentage of total team spinal load incurred by individual team members across the five LMDs were calculated. Paired t-tests were used to examine statistical differences between the gradients of the hand loads and spinal loads once the normality of the data was verified using the Kolmogorov-Smirnov test. Stepwise multiple regression was then used to determine the influence of the VHF, HHF, VHFma and HHFma on maximum and average L4/L5 torque, compression force, shear force and RPE. All statistical analysis was conducted using the Statistical Package for Social Sciences (SPSS version 10.0) and significance was accepted at $p < 0.05$. Descriptive statistics are reported as the mean \pm (SEM).

6.3 Results

6.3.1 Relationship between LMD and spinal loads and RPE

Significant positive correlations ($p < 0.05$) were detected for the relationship between LMD and all measures of maximum and average spinal load (i.e. maximum and average torque, compression force and shear force at L4/L5) at both the 30 kg and 60 kg load

conditions (Figure 6.2). Correlation coefficients ranged between 0.96 and 0.99 ($p < 0.01$) for maximum and average L4/L5 torque, between 0.91 and 0.99 ($p < 0.05$) for maximum and average L4/L5 compression force and between 0.86 and 0.99 ($p < 0.05$) for maximum and average L4/L5 shear force. RPE was also significantly correlated with LMD at both the 30 kg ($r = 0.98$, $p = 0.001$) and 60 kg ($r = 0.98$, $p = 0.002$) load conditions (Figure 6.2).

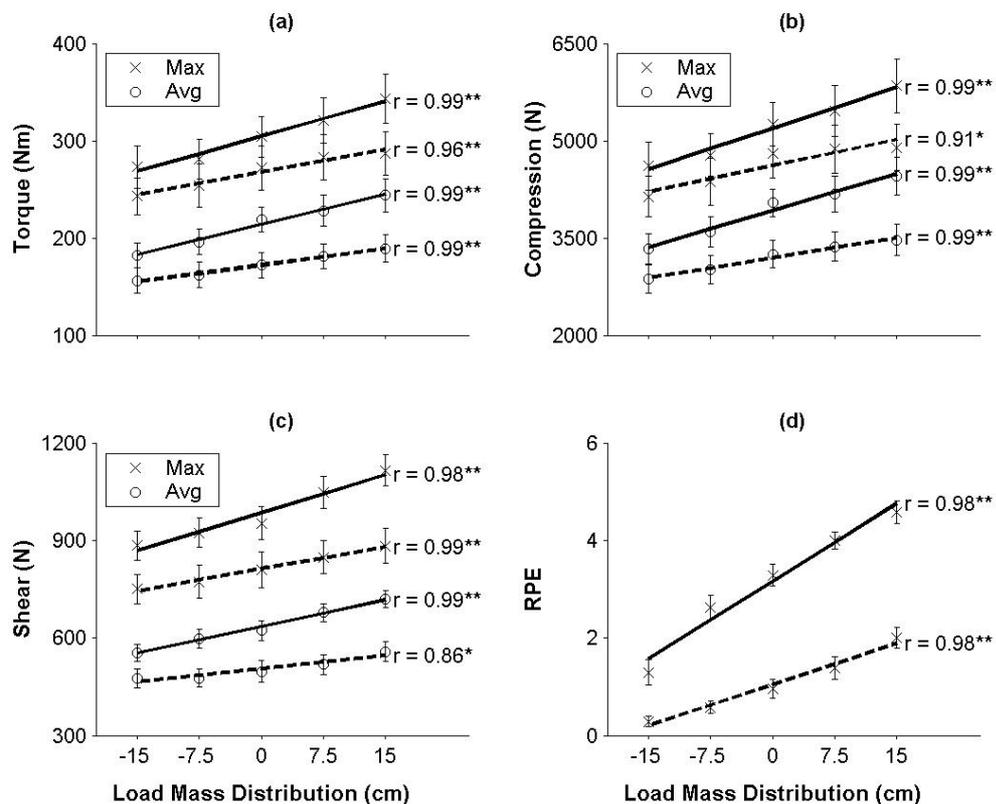


Figure 6.2: Relationship between load mass distribution and maximum (x) and average (o) L4/L5 torque, compression force, shear force and RPE at the 30 kg loading condition (dashed lines) and the 60 kg loading condition (solid lines). Pearson product moment correlations (r) and significance levels are also shown ($*$ = $p < 0.05$, $**$ = $p < 0.01$). A positive box mass distribution represents the heavy end of the load, and a negative box mass distribution represents the light end of the load.

6.3.2 Comparison of spinal loads with recommended loading limits

The maximum L4/L5 compression force experienced by the individual team members remained between the NIOSH Action Limit (AL) of 3433 N and Maximum Permissible Limit (MPL) of 6376 during all lift conditions (NIOSH, 1981). Similarly, the maximum L4/L5 shear force experienced under all lift conditions was between the corresponding shear limits (500 – 1000 N) proposed by McGill et al. (1998).

6.3.3 Relationship between LMD and VHF, HHF, VHFma and HHFma

Significant positive correlations were detected for the relationship between the maximum and average hand forces (VHF and HHF) and LMD at both the 30 kg and 60 kg loading conditions (Figure 6.3). The correlation between VHF and LMD for all conditions was 0.99 ($p < 0.01$), whilst the correlation between maximum and average HHF and LMD ranged from 0.86 to 0.95 ($p < 0.05$). Significant negative correlations were also found between maximum and average VHFma and LMD at the 30 kg ($r = -0.96$, $p < 0.01$) and 60 kg ($r = -0.88$ – -0.95 , $p < 0.05$) loading conditions.

6.3.4 Spinal load asymmetries during team lifts with unevenly balanced load

As expected the VHF was highly correlated with LMD ($r = 1.00$, $p = 0.001$, averaged across 30 kg and 60 kg load mass conditions). If LMD were the only factor to influence spinal loading during asymmetrical team lifts it would be expected that the spinal loads experienced by the team members at opposite ends of the box would be in the same ratio as the VHF ratios. However, the asymmetry in spinal loading variables was found to be less than the asymmetry in VHF across all asymmetric loading conditions. This finding is depicted in Figure 6.4, which shows that the slope of each spinal loading variable with respect to LMD was significantly less ($p < 0.01$) than the slope of the

VHF with respect to LMD. When the LMD ratio was 50:50 the ratio of the total spinal load experienced by each team member was also 50:50, when averaged across all loading conditions. However, at the 59:41 LMD ratio, maximum and average spinal loads were in the ratio 53:47 when averaged across the 30 kg and 60 kg loads. Similarly, at the 69:31 LMD ratio the maximum and average spinal loads were in the ratio 56:44 when averaged across the 30 kg and 60 kg loading conditions.

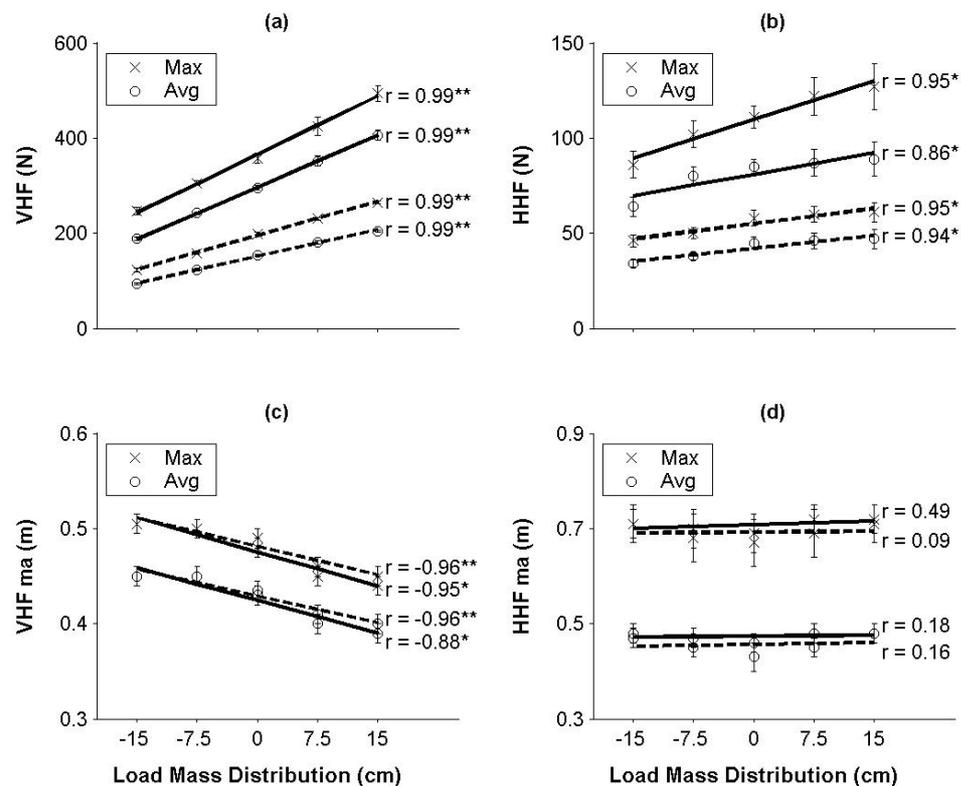


Figure 6.3: Relationship between load mass distribution and maximum (x) and average (o) horizontal and vertical hand forces (VHF and HHF) and their respective moment arms relative to L4/L5 (VHFma and HHFma) at the 30 kg loading condition (dashed lines) and the 60 kg loading condition (solid lines). Pearson product moment correlations (r) and significance levels are also shown ($*$ = $p < 0.05$, $**$ = $p < 0.01$). A positive box mass distribution represents the heavy end of the load, and a negative box mass distribution represents the light end of the load.

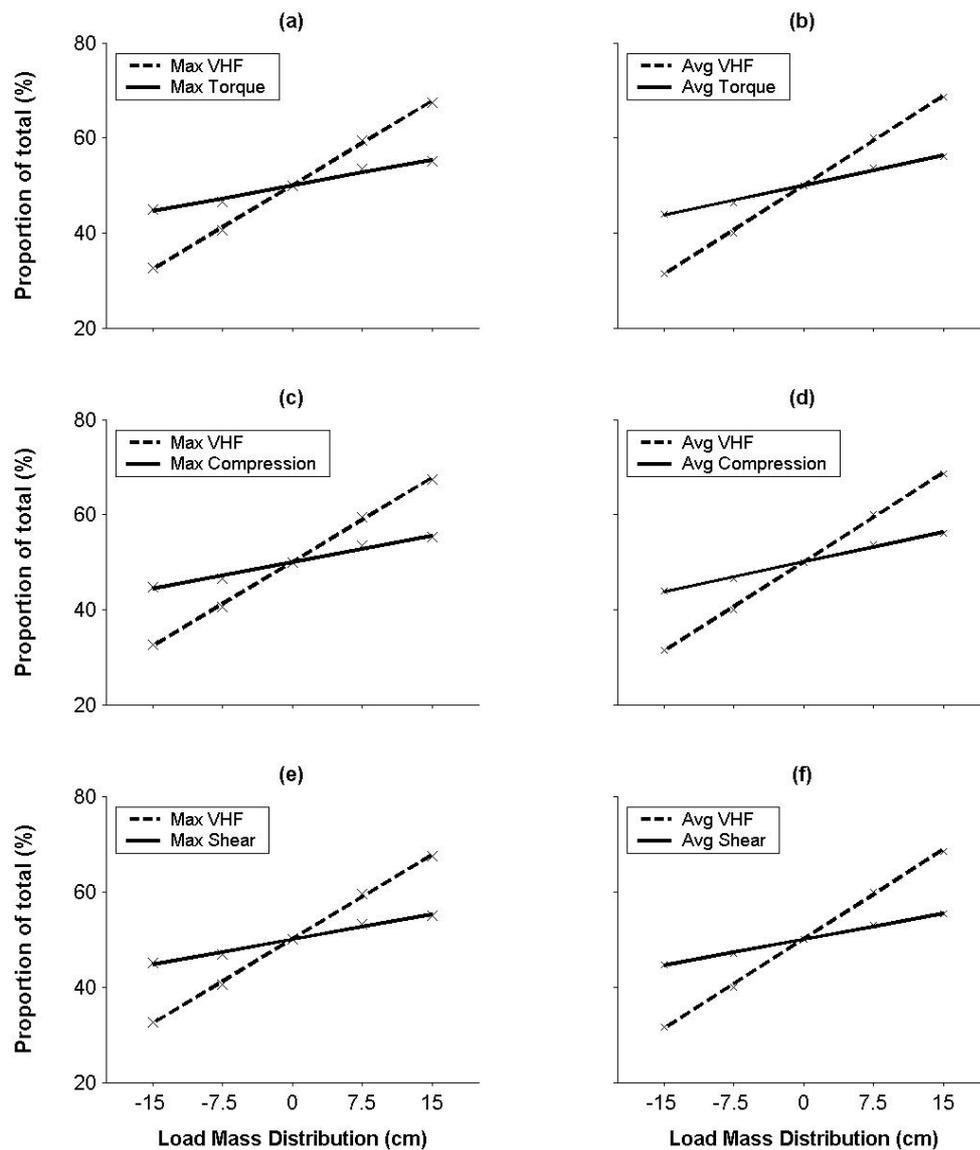


Figure 6.4: Regression lines showing the relationship between LMD and maximum vertical hand force and maximum spinal loads (a, c and e), and between LMD and average vertical hand force and average spinal loads (b, d and f). In each case the slope of the spinal loading curve is significantly less ($p < 0.001$) than the slope of the vertical hand force curve. A positive box mass distribution represents the heavy end of the load, and a negative box mass distribution represents the light end of the load.

6.3.5 Relationship between spinal loads and VHF, HHF, VHFma and HHFma

Pearson correlation coefficients indicating the relationship between maximum and average spinal loads (L4/L5 torque, compression force and shear force) and RPE, and maximum and average VHF, HHF, VHFma and HHFma are presented in Table 6.1. As expected, significant correlations were found between all indicators of spinal load and VHF ($r = 0.97 - 0.99$, $p < 0.01$). Significant correlations were also found between maximum and average spinal loads and both HHF ($r = 0.91 - 0.98$, $p < 0.05$) and VHFma ($r = -0.90 - -0.98$, $p < 0.05$), with the exception of average shear force which was not significantly correlated with average HHF ($r = 0.80$, $p = 0.106$). No significant correlations were found between HHFma and any of the indicators of spinal load.

Table 6.1: Correlation coefficients for the relationships between maximum and average L4/L5 torque, compression force, shear force and RPE and maximum and average VHF, HHF, VHFma and HHFma.

		Maximum			
		VHF (N)	HHF (N)	VHFma (m)	HHFma (m)
Maximum	L4/L5 torque (Nm)	0.99 **	0.95 *	-0.98 **	0.35
	L4/L5 compression (N)	0.97 **	0.96 **	-0.96 *	0.25
	L4/L5 shear (N)	0.99 **	0.92 *	-0.98 **	0.52
	RPE	0.99 **	0.93 *	-0.95 *	0.33
		Average			
		VHF (N)	HHF (N)	VHFma (m)	HHFma (m)
Average	L4/L5 torque (Nm)	0.99 **	0.91 *	-0.92 *	0.07
	L4/L5 compression (N)	0.99 **	0.93 *	-0.90 *	0.01
	L4/L5 shear (N)	0.97 **	0.80	-0.96 **	0.34
	RPE	0.99 **	0.98 **	-0.92 *	0.06

* = $p < 0.05$, ** = $p < 0.01$.

Stepwise linear regression was conducted to determine which variables were best able to predict spinal loads (i.e. maximum and average L4/L5 torque, compression force and shear force). The independent variables included in the regression analysis were the vertical and horizontal components of the hand forces (VHF and HHF) and their respective moment arms (VHFma and HHFma) according to Equation 6.1.

$$Y = A_0 + A_1(VHF) + A_2(HHF) + A_3(VHFma) + A_4(HHFma) \quad \text{Equation 6.1}$$

For Equation 1, Y is the dependent (i.e. spinal load) variable, VHF, HHF, VHFma and HHFma are the respective independent or predictor variables, and A_0 , A_1 , A_2 , A_3 , and A_4 are the regression coefficients. Regression coefficients for maximum spinal loads were determined using maximum hand force and moment arm data, whilst regression coefficients for average spinal loads were determined using average hand force and moment arm data. Regression analysis of the RPE scores was conducted using both maximum and average predictor variables.

Regression coefficients for predicting spinal loading variables as a function of VHF, HHF, VHFma and HHFma are presented in Table 6.2. Maximum and average L4/L5 torque and compression force were best predicted as a function of VHF alone ($R^2 = 0.95 - 0.98$). Maximum L4/L5 shear force was best predicted as a function of VHF, HHF and HHFma ($R^2 = 0.99$), whereas average L4/L5 shear force was best predicted using VHFma ($R^2 = 0.97$). The independent variables that best predicted RPE were average VHF and maximum VHFma ($R^2 = 0.99$).

Table 6.2: Stepwise multiple regression coefficients for predicting maximum and average L4/L5 torque, compression force, shear force and RPE collapsed across the 30 kg and 60 kg loading conditions (mean \pm SEM).

Y	A_0	A_1	A_2	A_3	A_4	R^2
Maximum Torque (Nm)	201 (8)	0.31 (0.03)	-	-	-	0.98
Average Torque (Nm)	128 (4)	0.29 (0.02)	-	-	-	0.99
Maximum Compression (N)	3424 (204)	5.28 (0.71)	-	-	-	0.95
Average Compression (N)	2367 (99)	5.32 (0.43)	-	-	-	0.98
Maximum Shear (N)	74 (4)	0.80 (0.01)	0.60 (0.18)	-	787 (4)	0.99
Average Shear (N)	1357 (77)	-	-	-1846 (179)	-	0.97
RPE	-7.5 (0.7)	0.02 (0.001)	-	11.5 (1.3)	-	0.99

$$Y = A_0 + A_1(VHF) + A_2(HHF) + A_3(VHFma) + A_4(HHFma)$$

6.4 Discussion

The purpose of this study was to determine the relationship between load mass distribution (LMD) and the spinal loads experienced by the individual team members during two-person team lifting tasks. Spinal loads were estimated using a simple biomechanical model of the lumbar spine. It was hypothesised that changing the distribution of the mass within the box would influence spinal loads. However, the magnitude of this effect was unknown because factors other than the distribution of the load mass had the potential to influence spinal loads during team lifting.

All measures of spinal load (i.e. maximum and average torque, compression force and shear force at L4/L5) were positively correlated with LMD. That is, spinal loads increased as the LMD ratio was increased from a minimum of 31% of the total load to a maximum of 69% of the total load. This finding was expected because increases in LMD resulted in corresponding increases in the effective mass handled by the team member, with obvious consequences for spinal loading. Similarly, the positive

correlation between RPE and LMD indicated that the lift was perceived to become more strenuous as LMD increased.

However, results of this study also indicated that the asymmetry in spinal loads between team members during unbalanced lifts were less than the asymmetry in hand loads. As LMD was systematically increased by approximately 9-10% of the total load being lifted (i.e. 31, 41, 50, 59 and 69%), the corresponding increase in spinal load was only 3% (i.e. 44, 47, 50, 53, 56%). This finding indicated that although spinal loads increased with LMD, this increase was less than that expected from the influence of LMD alone. It was therefore apparent that factors other than LMD were also contributing to the spinal loading patterns experienced by members of the two-person lifting team.

One factor that might explain the reduced asymmetry in spinal loads compared to hand loads during unbalanced lifts is the effect of body weight above the level of L4/L5. Because body weight above L4/L5 remains constant, a given increase in hand load results in a smaller percentage change in spinal load. To quantify this effect spinal loads associated with each LMD were calculated for the lifting posture depicted in Figure 6.1. Incremental increases in LMD of 9% resulted in spinal load increases of approximately 8%. It was therefore evident that body weight above L4/L5 had a relatively minor influence on spinal load asymmetry and that other factors must explain why a 9% increase in load mass was associated with only a 3% increase in spinal load. Subsequent analysis indicated that the two factors which minimised the increases in spinal load with increasing LMD were the horizontal “pulling” force acting on the hand and the moment arm of the load weight force acting through the hand with respect to the lumbar spine.

When the hand is below the level of the L4/L5 joint, the horizontal component of the hand force generates an extension torque about the lumbar spine that in turn reduces the extension torque required by the lumbar extensors (Dennis and Barrett, 2002). When either the load mass or LMD was increased to make the lift more demanding, a larger

pulling force was applied. As well as generating an extension torque about L4/L5, the larger pulling force exerted by the team member at the heavier end of the load caused the load to translate in the direction of that team member, thereby reducing the moment arm of the weight of the load. A reduction in the moment arm of the load weight reduced the flexion torque of the load about the lumbar spine and subsequently reduced the L4/L5 extension torque and compression force for that team member. This method for reducing spinal load (i.e. keeping the load close to the body) is well known and has a major influence on the weight that can be safely lifted by an individual (Waters et al., 1993; NIOSH, 1994). Although the horizontal pulling force and the moment arm of the vertical force acting through the hand were responsible for decreasing the asymmetry in spinal loads relative to the asymmetry in hand loads, they did not account for significantly more variance than the magnitude of the vertical hand force alone when predicting maximum and average L4/L5 torque and L4/L5 compression force using multiple regression.

The overall effect of the two-person team lifting coordination strategy observed in the present study was to decrease the asymmetry in spinal loading between lifting team members relative to the asymmetry in the LMD. The use of this coordination pattern seems to be an attempt to prevent the team member at the heavy end of the load from being subjected to excessive spinal loads. However, the minimisation of the spinal load incurred by the team member at the heavy end of the load occurs at the expense of the team member at the lighter end of the load, who experiences proportionally higher spinal loads than the proportion of the total load mass they lift.

6.5 Conclusion

Based on the results of this study it is concluded that asymmetries in load mass during two-person team lifting resulted in asymmetries in the lumbar spinal loads experienced by the respective team members. The distribution of mass within a load therefore needs to be considered when performing risk assessment of team lifting tasks. However, the asymmetry in spinal loads was less than the asymmetry in load mass. The two main factors responsible for this effect were the pulling force exerted on the load and the moment arm of the vertical force acting through the hands. When the effective mass experienced by a lifting team member was increased the pulling force applied to the box was increased which caused the box to translate towards that team member. Together, the increased pulling force and the reduced moment arm of the vertical hand force decreased the load acting on the spine for that team member, with a concomitant increase in the spinal load experienced by their lifting partner. This coordination strategy appears to be an attempt to protect the team member at the heavy end of the load from experiencing excessive spinal load.

Chapter 7

General Discussion

The purpose of this chapter is to synthesise and discuss the results obtained from the current study. A summary of the results from the three experiments is presented and subsequently synthesised to detail how these results expand upon the current body of knowledge on team lifting. From the synthesis of the results and taking the limitations of the study into account a set of recommendations for the use of team lifting in the workplace and recommendations for future research is made. Finally, the conclusions of the study are presented.

7.1 Summary of results

Most research to date has focused on assessing the spinal injury risks associated with individual rather than team lifting tasks. As a result, the guidelines governing the use of team lifting are based largely upon anecdotal rather than empirical evidence. Consequently, the three experiments conducted as part of this study were designed to assess the injury risks associated with team lifting tasks via the measurement of lumbar spinal loads. The first experiment compared the spinal loads during individual and team lifts to determine if the spinal injury risks were higher during team lifting than individual lifting, as is inferred by the Australian Manual Handling: National Code of Practice (NOHSC, 1990). In the second experiment, the necessity of team members to be of matched standing height, as suggested by the Manual Handling: National Code of Practice (NOHSC, 1990) was examined. The third experiment investigated the effect of the load mass distribution during team lifting on the relative spinal loads experienced by each team member. It was anticipated that the results of these studies would add to the limited body of knowledge on team lifting and thus aid in the development of guidelines related to the use of team lifting in the workplace.

Experiment 1: Spinal loads during individual and team lifting.

- Maximum and average L4/L5 torque and compression force in the sagittal plane were 20% lower for team compared to load-matched individual lifting tasks. Therefore, the null hypothesis that there is no significant difference in the spinal loads experienced during individual versus team lifting was rejected.
- The reduced spinal load experienced in team compared to individual lifts was due to an increased horizontal pulling force and the closer position of the hands relative to the body during team lifting.

Experiment 2: Spinal loads during two-person team lifting: effect of matched versus unmatched standing height.

- Maximum L4/L5 spinal loads were not significantly different between team lifting tasks performed with team members of matched or unmatched standing height. Therefore, the null hypothesis that there is no significant difference in the maximum spinal loads experienced by the individual team members of height matched versus height unmatched lifting teams was accepted.
- L4/L5 torque and compression force was significantly higher for the taller subjects and significant lower for the shorter subjects at the end of the unmatched compared to the matched lifting trials.

Experiment 3: Spinal loads during two-person team lifting: effect of load mass distribution.

- Changes in load mass distribution did not result in equivalent changes in the relevant distribution of spinal load between members of a two-person lifting team. Therefore, the null hypothesis that a change in the distribution of the load mass between the individuals of a lifting team results in an equivalent change in the distribution of spinal load between those team members was rejected.
- The minimisation of uneven spinal load distribution between team members occurred because the member at the heavier end of the load pulled the load closer to the body. Consequently, the increase HHF and decreased VHF_{ma} reduced the spinal load experienced by the team member at the heavier end of the load.

7.2 Synthesis of the results

Results from the present study indicate that sagittal plane spinal loads are lower for symmetrical lifting tasks performed by a two-person team rather than by an individual. This is expected in situations where the load mass is the same for both individual and team lifting tasks because the load lifted by each team member is effectively halved, as was the case for the research conducted by Mital and Motorwala (1995). However, in the current study and in the study conducted by Marras et al. (1999), the sagittal plane spinal loads during two-person team lifts were lower even though the load mass was twice as heavy in the team compared to the individual lifting tasks.

Marras et al. (1999) suggested that the lower spinal loads found in team compared to individual lifting tasks might be due to a change in the kinematic strategy that involves “less hip tilt and more trunk momentum”. However, the first experiment of the current study demonstrated that spinal loads in team lifting tasks was reduced primarily by the increased horizontal pulling force at the hands (HHF) and the ability of the individual team members to position the hands closer to the body (VHFma). An increase in the HHF and a decrease in the VHFma reduces the load on the lumbar spine by reducing the torque required by the spinal extensors. These two factors (increased HHF and reduced VHFma) were found to make approximately equal contributions to the reduction of sagittal spinal load found during team compared to individual lifts.

Although horizontal pulling forces (HHF) reduced spinal load in the current study, this only occurred because the hands were vertically positioned below the L4/L5 joint. In fact, during over head tasks a horizontal pulling force produces a flexion torque at L4/L5 that will increase the load on the lumbar spine (Hoozmans et al., 1998; de Looze, 2000). Furthermore, the extent to which the HHF is able to reduce spinal load depends not only on the magnitude of the HHF but also on the magnitude of the HHF moment arm (HHFma). Accordingly, during the floor to knuckle high lifting tasks performed in

the current study, the load relieving effects of the HHF were greatest at the start of the lift when the HHF_{ma} was a maximum, which was also when the largest spinal loads occurred. Another factor that must be considered when assessing the ability of the HHF to reduce spinal load is the increased risk that the hand may ‘slip off’ the load if the coefficient of friction between the hand and load is poor, or that the lifter may slip on the floor if the coefficient of friction between the feet and floor is low (Grieve, 1983; Lee et al., 1992). Finally, although the larger HHF in team lifts had no significant effect on the elbow or shoulder joint torques in the present study, the HHF associated with pushing and pulling tasks has been shown to increase the load experienced at joints other than the lumbar spine (Ciriello et al., 1993; Hoozmans et al., 1998). All of the effects of the HHF must be considered before the potential ability of the HHF to reduce the risks of spinal injury can be assessed.

Decreases in the VHF_{ma} that occurred during team compared to individual lifts in the present study were due to differences in the positions of the hands on the load. During individual lifts the hands were positioned in the middle of the load in order to maintain the balance of the box, whilst in team lifts subjects were able to position the hands at the ends of the load. Thus, if predesignated handle positions do not accommodate for team members to lift at the ends of the load, then the ability of the VHF_{ma} to reduce spinal load may be compromised. Indeed, Marras et al. (1999) found that if box handles were appropriately placed then spinal loads were reduced by the same amount as lowering the load by 4.5 kg. Whilst the current study investigated lifting tasks, it is possible that the effect of the VHF_{ma} on spinal load may be different during manual handling carrying tasks. During carrying tasks it is likely that team members will be able to step toward the load, which reduces the VHF_{ma} and can alter the manual handling technique used (Burgess-Limerick and Abernethy, 1998). During carrying tasks it is also possible that individuals may be better able to perform the task whilst the load is in contact with the body, effectively reducing the VHF_{ma} (Datta and Ramanathan, 1971), compared to individuals in a lifting team.

Therefore, the two factors that have arisen as critical in influencing spinal load in team lifting are the horizontal hand force (HHF) and the vertical hand force moment arm (VHFma). Both of these factors were consistently found in all three experiments of the current study to play a role in minimising spinal loads. Results indicate that an increased HHF and the minimisation of the VHFma are the primary ways in which spinal loads are reduced in team compared to individual lifts. In addition, the members of the lifting teams were able to coordinate the lifts in a manner that enabled the load relieving effects of the HHF and VHFma to favour the team member at increased risk of spinal injury.

The team lifting tasks assessed in the current study were performed in a coordinated manner that enabled the dynamic balance of the team to be maintained. As such, a perceived task constraint was to lift the box vertically whilst keeping the box in a stable horizontal and vertical position relative to each team member throughout the lift. A dynamic but stable position of the box relative to each team member is essential, as the balance of the team would be compromised if the position of the box became excessively skewed away from one team member.

The horizontal position of the box relative to each team member is controlled by the difference in HHF between team members. During team lifting tasks performed with height matched and evenly distributed loading conditions the team members generated equivalent pulling forces throughout the lift. As reported in the first experiment, the variability of the HHF between team members was less than half of the HHF variability between lifters in different teams. The ability of team members to match the HHF of their lifting partner is achieved via the mechanical coupling of the load between each lifter. However, when the load mass or relative team height was asymmetrical, then the team member at the greatest risk of spinal injury generated a larger HHF than their lifting partner to mitigate increases in spinal load.

During lifting tasks performed with matched team height and evenly distributed load the ensemble averages of the box incline indicated that the box remained horizontal throughout the duration of the lifting trials. Therefore, the vertical position of the box relative to each team member was kept constant throughout all lifting trials performed with matched team member heights and evenly distributed load mass. To do this the initial timing of the lift was coordinated so that each team member initiated the lift synchronously. In all three experiments the use of a verbal countdown resulted in a negligible time difference (0.02 s) in the initiation of the lift between team members. The dynamic coordination throughout the rest of the lift was possible via the mechanical hand-load coupling and visual information between lifting team members (Schmidt and Turvey, 1994). In contrast to the matched team lifting tasks, during the final stages of the unmatched lifting trials the box did not remain level and achieved a maximum incline of 5 degrees. However, even in this situation the team members coordinate the lift in an attempt to keep the box level by reducing the trunk extension of the taller team member at the end of the lift. If both team members lifted to a fully upright position it is expected that the box incline at the end of unmatched lifting trials would have been much greater than 5 degrees. Further investigation of the spatio-temporal information used both within an individual and between team members is required before definitive conclusions can be made about the inter-person information that is used to coordinate team lifting tasks (Paulignan et al., 1989; Turvey, 1990; Turvey et al., 1990).

Whilst the coordinative patterns of the HHF and VHFma between team members may not be the same during team lifting tasks with asymmetrical team member heights or loads, this uneven HHF and VHFma between team members represents a coordination pattern that mitigates the spinal load for the team member with the greatest risk of spinal injury. Therefore, rather than team lifting being a task that is performed by two independently exclusive individuals it appears that the interactive nature of team lifts

enables team members to work as a singular unit (Marsh et al., 2003), thus ensuring that no one team member incurs excessive spinal load.

All subjects used in the present study were novice lifters, with no experience or formal training in manual handling techniques. Despite the lack of training, lifting teams utilised the effects of the HHF and VHFma to reduce the loads on the lumbar spine. Thus, it appears novice lifters have an innate ability to coordinate the lifting task in order to reduce spinal load whilst still enabling the tasks to be effectively performed. Currently, it is unknown if experienced lifting teams can be trained to exploit the ability of the HHF and VHFma to reduce spinal load. However, at least one study by Chang et al. (2000) indicates that training and practice can result in individuals being able to perform precise pulling tasks that reduce lumbar torque during standing movements.

Although lifting teams were able to manipulate the HHF and VHFma to minimise excessive loads imposed upon the lumbar spine, the limitations of the study must be understood before recommendations with respect to the use of team lifting in the workplace can be made. Furthermore, factors that have not been examined by the current experimental research can also affect spinal load during team lifting and must be taken into account when assessing the injury risks associated with team lifting.

The lifting tasks analysed in the present study were all performed in the sagittal plane using 2-D analysis techniques. Asymmetrical lifting tasks can involve torsional forces that are not accounted for in 2-D analysis and may incur higher spinal loads during team than individual lifting tasks. Indeed, Marras et al. (1999) reported that lateral spinal loads were significantly higher for persons performing asymmetrical team lifts than asymmetrical individual lifts. However, Bone et al. (1990) indicated that asymmetrical lifting tasks analysed using 2-D models can be out of the sagittal plane by as much as 30 degrees before errors in spinal load estimates reach 10%. Thus, it should not be assumed

that the results obtained in the current study apply to lifting tasks with a significant asymmetrical component.

The spinal model used in the present study was a single equivalent muscle model. This type of model has been extensively used to compare the relative spinal loads during various manual handling tasks (Tsuang, 1992; Mital and Motorwala, 1995; Schipplein et al., 1995; Lavender et al., 2003). Although this model was validated against commercially available models (3D-SSPP and 4D-WATBAK), the reductionist nature of the current model reduces its capacity to produce absolute spinal load estimates that are accurate for all subjects. A single equivalent extensor force with a moment arm of 6 cm can result in spinal loads being inaccurate for particularly large or small subjects, and thus comparisons with the action and maximum permissible limits proposed by NIOSH (1981) may not be valid for these subjects.

The box that was used to measure hand loads in the current research was regularly shaped and provided good hand positioning on the load. However, there may be occupational team lifting situations where the quality of hand-load coupling is poor. A decrease in the quality of the hand-load coupling has previously been linked to decreased lifting capacity (Garg and Saxena, 1980; Smith and Jiang, 1984; Drury et al., 1989). Furthermore, in situations where the hand-load coupling is poor the chance of the load slipping in the hands is increased (Lee et al., 1992). Perturbations of the load may also have unwanted consequences in team lifting, where a team member may have to modify lifting technique to compensate for the actions of the other team member (Lee and Lee, 2002). The results from the present study may also not apply to situations where the load is not static (i.e. liquid contained within an object or lifting a person). Under these conditions the dynamic nature of the load requires constant adjustment by the team members. Thus, changes made by one team member to adjust to an unstable load may require sudden countermeasures by other team members that may involve excessive force or range of movement, which can increase the risk of spinal injury.

7.3 Recommendations for the use of team lifting

The current guidelines on team lifting (WHS, 2000; NOHSC, 1990) recommend that lifting teams should not be routinely used as the first option in occupational risk management. Further, if team lifting is to be used the guidelines advise the use of team members of matched height and build and that the lift should be coordinated by a team leader. Based on results from the present study there are several recommendations that can reduce the risk of spinal injury during two-person team lifting tasks and thus these findings can aid in the formulation of more comprehensive team lifting guidelines.

- Keep the load as close to the body as possible throughout the lift (NIOSH, 1991). If possible handle the load from the ends. Accordingly, objects that are to be lifted by teams should incorporate appropriately positioned handles into the design (Marras et al., 1999).
- When lifting a load with the hands positioned vertically below the level of the lumbar spine a horizontal pulling force can substantially reduce spinal load (Dennis and Barrett, 2002). However, the effects of an increased pulling force on the loads experienced at joints other than the lumbar spine and the possible increased risk of the hands slipping on the load and/or the feet slipping on the floor must also be taken into account (Ciriello et al., 1993; Hoozmans et al., 1998).
- Team members do not need to be matched for standing height as long as each team member lifts to a fully upright position. Therefore, team members should lift to a fully upright position at the end of the lift, as a flexed trunk will increase the load on the lumbar spine (Dennis and Barrett, 2003).
- Lifting an object with an unevenly distributed load will not produce the same degree of uneven spinal load between team members. However, the team member at the heavier end of the load will still experience greater spinal load and thus is at increased risk of injury, which must be taken into account when examining the risks associated with a particular team lifting task.

- A verbal countdown produced by an arbitrarily selected team leader adequately enables the subjects to coordinate the initial part of the lift (Dennis and Barrett, 2002).

7.4 Recommendations for future research on team lifting

In the absence of mechanical aids, team lifting may indeed be a useful manual handling strategy that can reduce the risk of spinal injury relative to individual lifts by reducing spinal load under certain lifting conditions. However, there are many other factors apart from those examined in the current study that may affect the ability of team lifting to reduce spinal load. These conditions must be investigated if all of the risks associated with team lifting are to be identified. Below is a list of further experimental research that is required before more definitive guidelines on team lifting can be developed.

- The effects of load carrying upon spinal loads during team lifting.
- The effects of ground surface incline upon spinal loads during team lifting.
- The effects of lifting an unknown load mass upon spinal load during team lifting.
- The effects of a perturbation applied to the load upon spinal load during team lifting.
- The effects of accuracy requirements when lowering a load onto a specified target upon spinal load during team lifting.
- The effects of a non-static load (e.g. liquid) upon spinal loads during team lifting.
- The effects of lifting training on spinal loads during team lifting.
- The effects of experience (i.e. expert vs novice lifters) upon spinal load during team lifting.
- The investigation of the spatio-temporal information that is used by members of a lifting team to coordinate team lifting tasks.
- Finally, epidemiological studies need to be established on team lifting to accurately correlate the prevalence rate of LBP with team lifting, and the effectiveness of team lifting to reduce the risks of injury based upon the criteria advised in section 7.3.

7.5 Conclusions

Overall, results from this study have demonstrated that: (i) the lifting strategy used by two-person teams is distinguished from individual lifts by a greater use of horizontal pulling forces applied to the load and a decreased distance between the load and the lumbar spine, (ii) both the horizontal pulling force and the position of the hands on the load in team lifting have a load relieving effect on the lumbar spine and (iii) two-person team lifts performed by team members of unmatched standing height and with asymmetrical load mass appear to be coordinated in a manner that partially mitigates the increased spinal load for the team member at increased risk of spinal injury.

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Appendix A

2D Static Spine Model

Model description

A 2D static spine model is included here to provide the user with the opportunity to independently assess the effect of model inputs such as body posture and magnitude and direction of hand loads on model outputs such as L4/L5 compression and shear forces. The 2D static spine model is based on the dynamic spinal model described in chapter 3 but with the influence of inertial forces removed. The 2D static spine model is therefore only appropriate for analysing manual handling tasks that are performed slowly and in the sagittal plane. The program's graphical user interface (GUI) is displayed in Figure A.1.

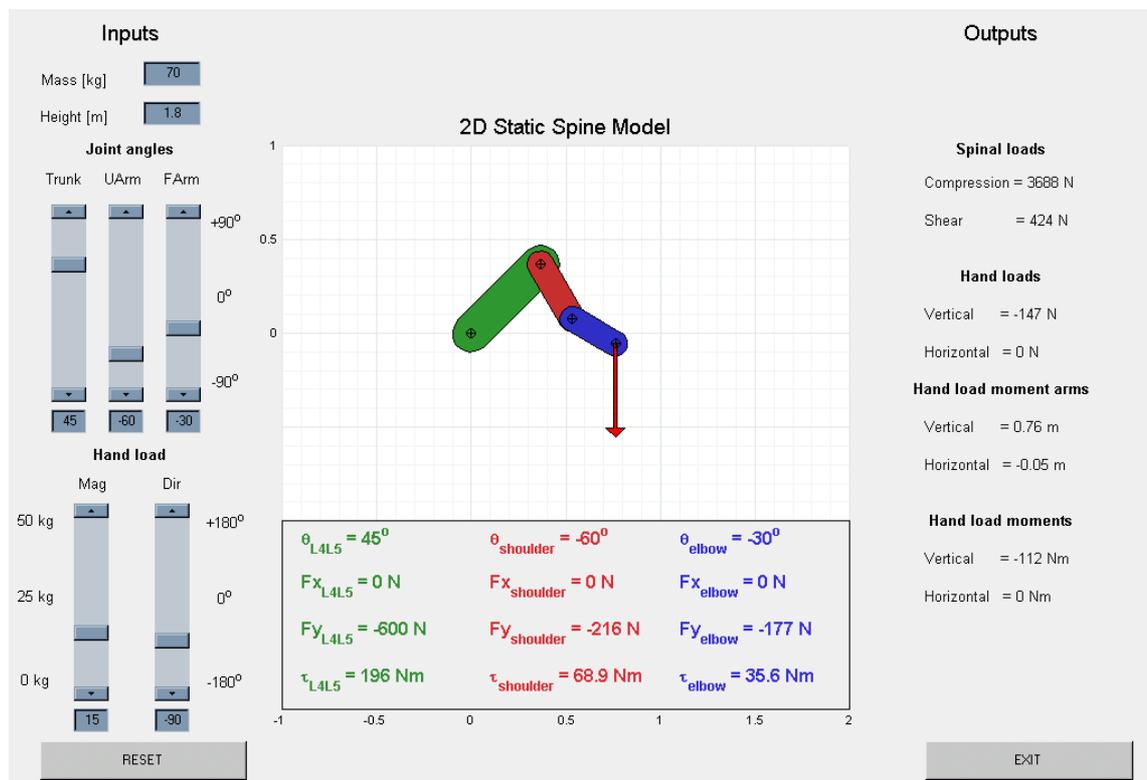


Figure A.1: Graphic user interface for the 2D static spine model.

Once the GUI has been activated the user may adjust the inputs on the left hand side of the GUI and observe the corresponding changes in model outputs displayed on the right

hand side of the GUI. A schematic representation of the trunk, upper arm and forearm segments as well as the force vector representing the hand load is displayed in the centre of the GUI. The model is fully interactive and outputs will instantaneously update in response to changes to any input parameters. Model inputs and outputs are listed in Table A.1.

Table A.1: Inputs and outputs of 2D static spine model.

Model inputs	Model outputs
<ul style="list-style-type: none"> • Body mass and height • Trunk, upper arm and forearm angles • Magnitude and direction of hand load 	<ul style="list-style-type: none"> • Elbow, shoulder and L4/L5 joint reaction forces and torques • L4/L5 compression and shear force • Vertical and horizontal hand forces • L4/L5 moments produced by vertical and horizontal hand forces

By using the model the effect of increasing the horizontal hand force or decreasing the moment arm of the vertical hand force on spinal loads may be easily demonstrated. A further simulation may also be performed to demonstrate the relationship between hand load and spinal load.

Installing and running the 2D static spine model

The 2D static spine model was developed using Matlab (version 6.0). Matlab version 6.0 or higher must be installed on your PC to run the model. The m-files required to run the 2D static spine model are provided on the CD contained in the back cover. To run the program, first copy the 3 files on the CD (spine.m, forkin.m and eqm.m) to a working directory on the hard disc drive of your PC then specify the path to the directory containing the 3 m-files in Matlab. The programs GUI is invoked by typing *spine* in the Matlab command window and then pressing the <return> key.