Experimental investigation and model-based simulation on lifting biomechanics : an emphasis on age effects

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2015

Song, J. (2015). Experimental investigation and model-based simulation on lifting biomechanics : an emphasis on age effects. Doctoral thesis, Nanyang Technological University, Singapore.

<https://hdl.handle.net/10356/62505>

<https://doi.org/10.32657/10356/62505>

**Experimental Investigation and Model-based Simulation
on Lifting Biomechanics: an Emphasis on Age Effects**

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**A Thesis Submitted to Nanyang Technological University in
Partial Fulfilment of the Requirements for the Degree of Doctor
of Philosophy**

2015

Acknowledgements

First, I would like to gratefully and sincerely thank my advisors, Dr. Qu Xingda and Dr. Chen Chun-Hsien, for their advice, guidance, encouragement, and most importantly their friendship during my graduate studies at Nanyang Technological University.

I would also thank to my best friends at Nanyang Technological University, Xinyao, Fui Ling, Sun Ke, Celine, Fitri, Zhiqing, Sam, Xiling, Xiaojing and Shuli. I cannot list all your names here, but thank you all for making my time full of fun. Further thanks to my parents and my wife for their love and care, without which this dissertation would not have been possible.

Abstract

Lifting tasks are closely associated with low back pain (LBP) which is a highly prevalent musculoskeletal disorder (MSD) and causes tremendous economic costs for the society. Also, Load lifting makes perturbations to postural stability, and thus is often associated with an increased incidence of falls. Therefore, examining lifting biomechanics and postural stability can help provide implications on ergonomic interventions including optimal lifting task designs and effective training programs, and eventually well address occupational LBP and fall injury problems related to manual lifting. In this dissertation, lifting biomechanics and postural stability were investigated using a combination of experimental studies and model-based simulation.

Experimental studies

In the experimental studies, age-related biomechanical differences were examined during symmetric and asymmetric lifting. Eleven younger and twelve older participants were instructed to perform lifting tasks defined by different combinations of destination positions and load weights. Lifting biomechanics was assessed using the whole-body kinematics and low back moment. Postural stability was examined using the inclination angles of the centre of mass (COM) and the centre of pressure (COP) as well as ground reaction forces.

The results regarding the lifting biomechanics showed that older adults adopted safer lifting strategies compared with younger adults in order to reduce low back load. For instance, the peak trunk sagittal flexion and transverse twisting angles were significantly lower (32% and 22% less, respectively) in older adults compared with those in younger ones. However, the average low back moment was significantly higher (about 32%) in older adults than that in younger ones, most probably due to the age-related increased body weight. Based on these findings, we recommend that physical exercise programs may be a more effective ergonomic

intervention for reducing the risks of low back pain (LBP) in lifting among older adults, compared with instructions of safe lifting strategies. As for younger workers, instructions of safe lifting strategies would be effective in LBP risk reduction.

The results on the lifting postural stability showed that older adults had smaller peak posterior and leftward ground reaction forces than did younger adults (56.5% and 38.6% less, respectively), but there was no age-related differences in the peak COM-COP inclination angles. It was also found that asymmetric lifting and larger load weights were associated with poorer postural stability. Based on the findings, it might be suggested that in order to maintain postural stability during lifting, especially in older adults, the lifted load weight be adjusted according to individual strengths, and asymmetric lifting be avoided if possible.

Lifting motion simulation

In the model-based simulation, two hybrid models using single- and multi-objective optimization, respectively, are developed for lifting motion simulation. The human body is modelled using a 2-D five-segment model. The lifting motions are predicted by solving a non-linear optimization problem subject to various kinematic and kinetic constraints. Specifically, the joint angular velocities are bounded by time-dependent constraints constructed from actual motions. In the single-objective optimization model, the objective function is defined based on a minimum-physical-effort performance criterion. In the multi-objective optimization (MOO) model, two performance criteria, including minimum physical effort and maximum load motion smoothness, were optimized simultaneously using a weighted-sum MOO approach.

Symmetric lifting motions performed by younger and older adults under varied task conditions were simulated. Comparisons between the simulated motions and their corresponding actual motions were made to evaluate both models. The results showed that

the absolute joint angle errors were less than 10° , which suggests the proposed model is able to accurately simulate 2-D lifting motions. The proposed models are also comparable with the existing motion simulation models in terms of the prediction accuracy. Moreover, the prediction accuracy is improved for both age groups (6.3% - 28.3% decreases in prediction errors) after using the MOO approach, compared to the single-objective optimization approach. Therefore, the MOO approach is recommended as a superior method for 2-D lifting motion simulation compared with the single-objective optimization.

In summary, age-related differences in lifting biomechanics have been examined in experimental and model-based studies. The experimental results show that, compared with younger adults, older adults tend to use safer lifting strategies (e.g. reduced trunk flexion and lifting velocity) to reduce LBP risks and enhance postural stability during lifting. However, older adults may still have higher risks of LBP due to their relative larger low back load. Two hybrid models were developed for lifting motion simulation using the single-objective optimization and multi-optimization approaches, respectively. The models are able to predict accurate symmetric lifting motions for both younger and older adults, and reveal age-related differences in the mechanisms for planning lifting motions.

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Chapter 1 Introduction

1.1 Research Motivations

Musculoskeletal disorders (MSDs) are serious public health problems which are highly prevalent in various occupations (Alexopoulos et al. 2004; Fernandes et al. 2011; Habibi et al. 2008; Hsu et al. 2011). MSDs also induce enormous economic costs. In the United States, 46.4% of the population reported MSDs annually in the years of 2002 to 2004, and the total annual costs caused by MSDs reached almost \$950 billion in 2006 (American Academy of Orthopaedic Surgeons 2011).

The high prevalence and tremendous economic costs of MSDs have motivated a variety of research to identify risk factors (i.e. causes) leading to MSDs. These factors could be work-related, such as postures/motions, localized mechanical stresses and vibrations (Kilbom et al. 1996; Passier and McPhail 2011), or personal factors, such as gender, age and obesity (Faucett and Werner 1999). In order to develop effective interventions for reducing risks of MSDs, there is a need to use biomechanical measures (e.g. electromyogram of skeletal muscles, human motions and joint loadings) to facilitate better understanding the associations of these factors with MSDs risks (Coenen et al. 2013; De Luca 1997; Granata and Wilson 2001). Two distinctive methods are often used to address this need (Robertson et al. 2013).

The first method is reactive and experiment-based, in which realistic biomechanical measures are measured when subjects are performing occupational tasks in laboratories or fields. The associations between the risk factors and MSDs in different working scenarios can be investigated by manipulating the task scenarios and subject attributes. The biomechanical measures collected using the experiment-based method is realistic and reliable, which is the key advantage of this method. However, this method is time-consuming and costly, and not

all working scenarios can be simulated in experiments. Moreover, the results in the experiments cannot be well generalized to novel situations due to experimental controls.

The second method is proactive and model-based, in which biomechanical measures (e.g. human motions, external and internal joint loads) are predicted using digital human models for different working scenarios (Abdel-Malek et al. 2006; Chaffin et al. 2007). Unlike the experiment-based method, the model-based method can predict human performances in a virtual environment without constructing the physical mockup. Thus, the model-based method is associated with higher efficiency and low cost. At the same time, use of model-based methods also allows minimizing the risk of exposing workers to ergonomic stressors and fatigue during work. More importantly, the model-based method is able to predict human performances in novel scenarios.

Low back pain (LBP) is one of the most prevalent and costly MSDs. In the US, nearly two-thirds of adults are affected by LBP during their life time (Lawrence et al. 1998), and the corresponding annual costs exceed \$100 billion (Katz 2006). Manual lifting is one important risk factor of LBP in occupations (Hoogendoorn et al. 2000; Kuiper et al. 2005). During lifting tasks, workers always have to bend, twist and stretch to reach and carry heavy loads while maintaining balance. Hence, lifting often induce high energy expenditure and spinal loadings, and consequently expose the workers to a high risk of LBP. Knowledge of lifting biomechanics can help specify design guidelines for ergonomic interventions for addressing the occupational LBP problems caused by manual lifting can be well addressed.

Although a great deal of research has been conducted to study lifting biomechanics, the optimal lifting strategy for reducing low back pain incidence is still unknown (van Dieën et al. 1999; Burgess-Limerick 2003). The effects of some risk factors (e.g. lifters' age) on lifting biomechanics and risks of LBP have not been well understood. Experimental biomechanics

studies could clarify the effects of these factors on LBP risks during lifting, and help propose better lifting strategies for various working scenarios and different populations. In addition, model-based biomechanics studies, especially lifting motion simulation, can predict lifting postures and dynamic motions, and further evaluate LBP risks in novel scenarios. Actually, motions and postures have been widely recognized as a primary risk factor for LBP (Caboer et al. 2000; Davis and Marras 2000a; Granata and Wilson 2001; Vingård et al. 2000). However, current motion prediction models are far from perfect. Human motions and postures, especially in novel scenarios, have not been well modelled (Chaffin 2005). Therefore, there is a need for exploring lifting biomechanics by further conducting lab-based experiments and developing better lifting motion prediction models.

1.2 Research Scope

In order to reveal age effects on lifting biomechanics, both experimental studies and model simulations were conducted. In the experimental studies, age-related changes in lifting biomechanics were examined under varied task conditions defined by different levels of load weights, destination height and task asymmetry. In the model simulation, two novel hybrid models for lifting motion simulation were developed using single-objective and multi-objective optimization approaches, respectively.

1.3 Specific Aims and Hypothesis

The specific aims of the experimental study are:

1. To examine the effects of the personal factor of interest (i.e., lifters' age) on lifting biomechanics;

2. To investigate the interaction effects between age and the work-related factors (i.e., load weight and destination height) on lifting biomechanics.

Their corresponding hypotheses are:

1. There would be significant age-related differences in lifting biomechanics;
2. There would be significant interaction effects between age and the selected work-related factors (i.e. load weight and destination height) on lifting biomechanics.

The objective of the model-based study is to develop and validate a novel lifting motion prediction model that can generate accurate lifting motions which account for the effects of task scenarios and individual differences. The corresponding hypothesis is: the proposed model is able to predict lifting motions more accurately than the existing models.

1.4 Organization of the Thesis

This dissertation is divided into nine chapters. Chapter 1 contains research motivations and objectives. In Chapter 2, literature on lifting biomechanics studies and human motion prediction is reviewed. Chapter 3 presents an experiment involving both symmetric and asymmetric lifting. In Chapter 4 and Chapter 5, the effects of work-related and personal factors (i.e. load weight, lifting destination height and age) on lifting biomechanics in symmetric and asymmetric lifting are presented and discussed, respectively. Chapter 6 presents the age-related differences in postural stability during lifting tasks. In Chapter 7 and 8, novel lifting motion prediction models are developed and validated. These models are used to provide information on age-related differences in the mechanisms of lifting motion planning. Chapter 9 summarizes the key findings and limitations of this dissertation research, and proposes future research problem.

Chapter 2 Literature Review

2.1 Low Back Pain and Manual Lifting

Low back pain (LBP) is a musculoskeletal disorder (MSD) that affects the spine and back tissues, and results in discomfort, pain and even disability. LBP is highly prevalent: about two-thirds of the adults have LBP at some time during their lives (Lawrence et al. 1998). It has been reported that, in the US, the lifetime prevalence of LBP (lasting more than 2 weeks) reached 13.8% in 1987 (Deyo and Helmick 1987), and the prevalence of chronic LBP rose from 3.9% in 1992 to 10.2% in 2006 (Freburger et al. 2009). Besides the US, high lifetime prevalence of LBP has also been reported in many other countries and districts, such as 70% in Denmark (Harreby et al. 1996), 59% in UK (Hillman et al. 1996) and 39% in Hong Kong (Lau et al. 1995). Worldwide, it is estimated that 37% of LBP are caused by occupational factors (Punnett et al. 2005). In the occupations requiring high physical effort (e.g. farmers, operators and service workers), the prevalence of LBP is even higher (Guo et al. 1999; Leigh and Sheetz 1989).

In addition to its high prevalence, LBP has brought tremendous economic cost worldwide. The total cost is comprised of two components: direct and indirect costs. The direct cost includes cost of medical treatments and health care; the indirect cost is mainly caused by work absences and decreased productivity of the patients. The annual cost of LBP is estimated to exceed \$100 billion in the US (Katz 2006). In many other countries, such as Australia (Walker et al. 2004), Japan (Shinohara et al. 1998) and European countries (Juniper et al. 2009; Maniadakis and Gray 2000; van Zundert and van Kleef 2005), enormous financial cost caused by LBP has also been reported.

Manual lifting is one of the most common manual tasks in various occupations, such as farm labourers, construction workers and health care workers. Previous epidemiological studies

have reported a close association of lifting with high incidence of work-related LBP (Engels et al. 1996; Frymoyer et al. 1983; Hoogendoorn et al. 2002; Prado-Leon et al. 2005). During lifting tasks, workers need to bend and twist their trunks, stretch their upper extremities, and carry heavy loads, etc. All of these actions may require high physical efforts, produce considerable spine loads, and consequently increase the risk of LBP. Moreover, long repetitive lifting tasks in industry tend to induce fatigue and make the human body more vulnerable to low back injuries.

Given the considerable costs of LBP and its close association with manual lifting, ergonomic interventions have been proposed to reduce the exposure of workers to the risk of LBP (Burton 2005). In order that the interventions could be effective, experiment-based and model-based studies should be conducted to examine the effects of various work-related and personal factors on lifting biomechanics and to evaluate the risk of LBP for the current workplace setups.

2.2 Experiment-based Studies on Lifting Biomechanics

Extensive studies have been carried out to understand and quantify the effects of risk factors on lifting biomechanics. Generally, the factors of interest in ergonomic study on the lifting task can be classified into two categories: work-related factors and personal factors.

2.2.1 Work-related Factors

Work-related factors are defined by the characteristics of the lifting task. They are closely related with task requirements and the workplace design, and therefore, frequently used in ergonomic interventions. Investigations of the effects of work-related factors on lifting biomechanics are summarized in this section.

2.2.1.1 Lifting Styles

Lifting styles, also called as lifting strategies, lifting techniques, etc., refer to the general kinematic features of lifting motions. The lifting style can be defined by the initial lifting posture. Two predominant styles, squat and stoop, are often observed in industrial work and experiments (Hsiang et al. 1997). The squat style is characterized with a minor trunk flexion but a large knee flexion in the starting posture. In contrast, the stoop style begins with a substantial trunk flexion but little or even no knee flexion. Besides these two extreme styles, workers also adopt lifting styles intermediately between them, such as semi-squat strategy (Burgess-Limerick 2003; Burgess-Limerick and Abernethy 1997a), which reflects a continuum nature of the lifting style. Burgess-Limerick and Abernethy (1997b) presented a postural index to quantify this continuum nature. The postural index was defined as the ratio between knee flexion and the sum of ankle, hip and lumbar vertebral flexion from normal standing in the initial lifting posture. According to this definition, the value of the index is about 0.11 for the stoop style and close to 0.80 for the squat style.

However, the definition of the lifting style in terms of the initial lifting posture cannot fully reflect the kinematic features of the lifting task because this static definition neglects the dynamic features of the lifting motion. Kjellberg et al. (1998) proposed a new concept regarding lifting styles that consists of two parts: the method of carrying out the task and the individual performance during the task. The first component, the ‘method’, was defined by joint ranges of motions, such as the trunk motion ranges which differentiate the squat and stoop lifting styles. The ‘individual performance’ referred to the dynamic properties of the motion, such as joint angular velocities and accelerations. To sufficiently account for lifting styles, the dynamic properties of lifting motions should be considered as well.

There is no consensus on which lifting style is associated with the minimum risk of injuries (Burgess-Limerick 2003; Hsiang et al. 1997; van Dieën et al. 1999). Normally, the squat style is considered to be better than the stoop style since the former one produces lower low-back muscle forces and internal spinal loads (Bazrgari et al. 2007). However, some disadvantages are also related with the squat style, such as reduced postural stability, high inspiratory ventilation and oxygen consumption (Burgess-Limerick 2003; Hagen et al. 1993, 1994; Hsiang et al. 1997). In fact, because occupational lifting tasks involved various working scenarios and people of different populations, there is unlikely to be a single best lifting style appropriate for all working conditions. For instance, in the free-style lifting task, subjects preferred to adopt the semi-squat style which enhanced inter-joint coordination and reduced the muscular effort (Burgess-Limerick 2003). It was also found that the free-style lifting technique generated the highest lifting capacity compared with both the squat and stoop techniques (Kuijer et al. 2012). In another study in which participants lifted very large-size loads, a novel lifting style, characterized by a wide foot placement, moderate knee flexion and a straight but not upright trunk, yielded 20% lower compression forces at the L5/S1 joint than did the stoop and squat styles (Kingma et al. 2010).

2.2.1.2 Lifting Velocity and Lifting Frequency

Lifting frequency and lifting velocity are two related factors in the manual handling task. Lifting frequency usually reflects task requirements. For instance, when workers lift weights from an assembly line, the velocity setting of the assembly line determines the lifting frequency. Lifting velocity, however, is the combined result of task requirements and individual selections. With a high lifting frequency requirement, the lifting velocity is likely to be high otherwise workers cannot follow the working progress. When the lifting frequency is low, the lifting velocity is mainly dependent on workers' personal selections.

Previous researches have reported that the inertial effects should not be negligible when estimating the spinal loads (Bushjoseph et al. 1988; Lindbeck and Arborelius 1991). Higher velocities/frequencies increase the spinal forces and decrease trunk muscle strength, and therefore increase the risk of LBP (Davis and Marras 2000a). Therefore, high velocities and frequencies are considered as the risk factors for low back injuries.

Hagen et al. (1995) studied the effects of lifting frequency on the thigh and lower-trunk motions in symmetric repetitive lifting. Participants were instructed to adopt both the squat and stoop strategies at two different frequencies of 10 and 20 lifts/min. The motion ranges of the angular displacements of the thigh and lower-trunk were recorded. When using stoop lifting, the lifting frequency had no significant effects on the motion ranges. However, when using squat lifting, the thigh motion ranges significantly decreased by 7.4° as the lifting frequency increased from 10 lifts/min to 20 lifts/min. This change might be the result of the quadriceps muscle fatigue in the repetitive lifting at the high lifting frequency. Therefore, high lifting frequency might be a potent stimulus for fatigue in repetitive lifting tasks.

Mirka and Kelaher (1995) investigated the effects of lifting frequency on trunk kinematics. Participants were required to transfer loads between two conveyers at different lifting frequencies (3, 6 and 9 lifts/min). Increased lifting frequencies were found to induce higher trunk sagittal angular accelerations ($p < 0.001$). Dai et al. (2010) investigated the effects of lifting frequency and horizontal load speed on trunk kinematics and muscle activities when lifting loads from a conveyor to a table. Four horizontal load speeds (0 m/s, 0.7 m/s, 1.3 m/s and 2.4 m/s) and two lifting frequencies (10 and 20 lifts/min) were involved in the experiment. Results revealed significantly increased trunk muscle co-activation with the higher load speeds and lifting frequency. It was also found that increased load speed from 0 m/s to 2.4 m/s resulted in significant increases in the trunk sagittal flexion (73° vs. 81°), but significant decreases in trunk sagittal ($480^{\circ}/s^2$ vs. $4^{\circ}/s^2$) and transverse ($200^{\circ}/s^2$ vs. $140^{\circ}/s^2$)

angular accelerations. It was argued that participants applied the load speed to reduce the trunk dynamics at a cost of increased trunk flexion and muscle activities.

Chen (2000) examined the effects of lifting frequency in a study determining the maximum acceptable weight of lift (MAWL) for Chinese males. Participants lifted the loads in three frequencies: one-time maximum (OTM), 1 lift/min, and 4 lifts/min. The OTM was considered as the infrequent task and 4 lifts/min was the frequent task. The MAWL decreased with the increased frequency. The analysis on the initial lifting posture showed that a more stoop technique was adopted in the frequent task because repetitive stoop-lifting was metabolically less demanding than squat-lifting. In infrequent tasks with heavier loads, the participants tended to increase the trunk vertical acceleration and at the load-landing phase, flexed their elbows and shoulders, and leaned their bodies forward. The lifting pattern at the load-landing phase helped shorten the moment arm about low back joint and compensate the weakness of the upper extremities. This lifting pattern in infrequent tasks indicated the importance of considering the upper extremity strength when designing the lifting tasks with heavy loads.

Zhang et al. (2003) studied the effects of lifting velocity on lumbar vertebral kinematics. The participants lifted the loads at two different speeds: a 'normal' self-paced speed and a 'faster-than-normal' motivated speed. The bony landmarks of seven spinous processes, C7, T7 and L2-S1, were placed with reflective makers. With this marker placement, the thoracic spine could be modeled with two rigid segments (i.e. C7-T6, T7-L1), and the movements of each vertebrae of the lumbar (L2-L5) spine could be recorded. The result showed that, the angular accelerations of all lumbar vertebrae (L2-L5) at the early phase of lifting were higher in high-velocity lifting trials, which might induce higher spinal loads.

Greenland et al. (2013) examined the effects of lifting velocity on cumulative and peak low back loads during symmetric lifting. In the experiment, participants lifted loads of two different weights (2.25 and 9 kg) at three different velocities (0.34, 0.44 and 0.64 m/s). Three lifting ranges were involved, including floor-to-shoulder, floor-to-waist and waist-to-shoulder. The results showed that increased lifting speed induced significantly higher peak low back load ($p < 0.001$), with a mean difference of 20% between fast (0.64m/s) and slow (0.34m/s) lifts. However, due to the longer lifting duration in the slow velocity condition, the cumulative low back load was significantly increased with decreased lifting velocities, with mean differences $> 80\%$ from the slow to fast condition. Therefore, besides the peak low back loads, the lifting duration should be also considered when determining risks of LBP during lifting.

2.2.1.3 Load Positions in the Workplaces

The initial load position and its final target have significant effects on lifting motions and spine loads, and may increase the risk of low back disorders if they are inappropriately located. In this section, studies on the effects of the initial horizontal distance of load from the human body, the lifting height, and the asymmetry of the load position on lifting biomechanics are summarized.

Burgess-Limerick and Abernethy (1998) investigated the effects of the initial horizontal load distance on the initial lifting posture. Participants lifted loads of different weights (3, 6 and 9 kg) from a starting height of 18 cm above the floor in the sagittal plane. The initial horizontal distance was set at 20, 40 and 60cm. The results showed that the initial lifting posture was independent of the load weight, but largely influenced by the initial distance. The hip, knee and ankle flexion was found to be reduced by around 5° , 35° and 15° , respectively, as the ankle-load distance decreased from 60cm to 20cm, while lumbar vertebral flexion remained

relatively unchanged. Given that the larger horizontal distance produces larger load moments and spinal loads, reducing the distance should be considered in the lifting task design.

In realistic working conditions, workers are likely to shift the distant load towards them before they lift the load, so the effects of the load distance may be minimized. Therefore, Faber et al. (2011) explored how the effects of the initial horizontal load position on the peak L5/S1 total moment (PTM) were affected by task types (lab-based lifting and realistic lifting). The participants lifted loads from two initial horizontal positions (far and close) in three conditions: 1) the laboratory condition in which the load was lifted without transportation; 2) the realistic condition in which the load could be moved closer before lifted and 3) the familiarity condition in which the participants were familiar with the move-closer strategy after practice. No significant difference was found in the PTMs between the laboratory condition and the realistic condition when the participants lifted far-positioned loads. The PTMs for far-positioned load were 10% lower in the familiarity condition than those in the laboratory condition. PTMs for far- and close-positioned loads in the familiarity condition were not significantly different. Therefore, if the workers are allowed to shift the box before lift and familiar with this move-closer strategy, the effects of the initial horizontal load position on the PTMs will be minimized.

In addition to the horizontal load distance, the initial and final load vertical positions also affect lifting biomechanics. Hoozemans et al. (2008) quantified the effects of the initial load vertical position on the peak low back load. The participants symmetrically lifted loads with two weights (7.5kg and 15 kg) from four heights (from 0.32m to 1.55m) to the waist height. A regression model was developed to quantify the effects of the load weight and the initial lifting height on the low back moment and the spinal compression force. The result indicated that either increases in the lifting height or decreases in the lifting weight resulted in decreases in the moment and the compression force. Moreover, the regression results showed

a larger effect of the lifting height on the spinal loads than that of the load weight, so it was advised that increasing the initial lifting height was a more effective way to reduce the low back loads than decreasing the load weight.

Faber et al. (2009a) examined the effects of lifting height on low back and shoulder joint loadings during masonry work. Participants first lifted a building block, walked a few steps, and then placed the block onto a table. The initial lifting height in the block-lifting task was set at 10 and 90 cm, and the final lifting height in the block-placing task was set at heights of floor, iliac crest and shoulder levels. As found by Hoozemans et al. (2008), it was also found here that increasing lifting height from 10cm to 90cm was effective in reducing the low back load by 30 Nm. However, the shoulder lifting height resulted in relatively high shoulder loading. Therefore, it was suggested that the iliac crest lifting height was the optimal workplace setting.

In workplaces, workers are often required to lift loads outside the sagittal plane. Davis and Marras (2005) investigated how the positions of the load origin and destination impacted three-dimensional spine loads at L5/S1. Three height levels (elbow, knee and shoulder) and three asymmetric levels (60° clockwise, sagittally symmetric and 60° counter-clockwise) were used for both the load origin and destination positions. The starting and destination heights as well as starting task asymmetry had significant effects on the spine loads. It was found that the neutral lifting in which the initial and final load positions were both at the elbow height within the sagittal plane resulted in the lowest back load. Compared with the neutral lifting, increases of the peak values of the spine compression force (between 400 and 1900N), lateral shear force (100 to 150N) and anterior-posterior force (up to 400N) were observed in other lifting conditions. Therefore, lifting heights below the knee and above the shoulder as well as large asymmetry in the load position settings are associated with high risk of spinal injuries, and should be avoided in the workplace design.

Faber et al. (2009b) examined the low load loading in two conditions: lifting one load in front of the body and lifting two loads beside the body. Participants lifted a load weighted 20 kg in front of the body (single-load condition), and then lifted two 10-kg loads besides the body (split-load condition). Four lifting techniques (stoop, squat, straddle and kneeling) were involved in the experiment. The low back load was evaluated using net moments and spinal forces at the L5/S1 joint. The results showed significant decreases (8-32% in most cases) in the peak L5/S1 compression force in the split-load condition than that in the single-load condition. The reduction in low back loads could be attributed to the reduced load-L5/S1 distance in the split-load condition.

2.2.1.4 Load Weight

It has been reported that load weight significantly affects lifting biomechanics, and heavy load is always considered as a risk factor for LBP (Radwin and Lavender 1999). Many studies have been carried out to investigate how the increased load magnitude influences lifting biomechanics, including low back load, lifting kinematics, postural stability and muscle activities.

Davis and Marras (2000b) investigated the effects of increased load weight on trunk kinematics and spinal loads. Participants were instructed to symmetrically lift the load with varied weights (9.1 kg - 41.7 kg) from knee to elbow height. The result showed no significant increase in the spinal loads with small increases in load weight (3-9 kg) when the weight was small (< 18 kg) because the increased weights were offset by the decreased trunk sagittal velocity. However, when the weight was above 25 kg, the spinal loads increased significantly with even small increases in load weight (e.g., 19.6% increases in spinal compression forces as load weight increases from 20kg to 30kg). In another study, Hoozemans et al. (2008) quantified the combined effects of lifting height and load weight on the peak low back load

with a regression model. The result indicated that the spinal compression force at L5/S1 decreased significantly with the decreased load weight. These two studies indicated that heavy loads lead increased risk of low back injuries.

Changes in lifting kinematics with varied load weights have also been studied. Hagen et al. (1995) investigated the effects of load weight on the motion ranges of the thigh and trunk angular displacements using two lifting styles: squat and stoop. No effect of increased load weights on the two motion ranges was found in both styles. Zhang et al. (2003) studied the trunk kinematics with varied load weights (2.3 - 11.5 kg) in the symmetric lifting task. The spine was modelled not as a single rigid segment but several connected segments according to the spine's anatomical structure. The results showed a higher load increased the peak angular velocity and acceleration of the L5/S1 spine joint, which might increase the spinal loads.

The increased load weight also influences the coordination patterns between joints and body segments during lifting. Scholz (1993) studied the changes of coordination patterns in terms of relative timing between joint angles with varied load weights in a symmetric squat lifting task. The maximum lifting capacity (MLC) of each participant was measured prior to the lifting task, and the load weights were set at six equally spaced levels from 0 to 75% of each participant's MLC. It was found that the relative timing in the joint couples of ankle-hip, knee-back, ankle-back, knee-ankle and hip-back changed significantly in a linear manner with the increased load weight. In another study, Scholz et al. (1995) used the relative phase between joint angles (Burgess-limerick et al. 1993) to characterize the coordination patterns in the symmetric squat lifting. Participants lifted the load with varied weights from 15 to 75% of the individual's MLC. It was found that the relative phase between joint angles (i.e. knee and lumbar spine) increased in a quasi-linear manner during lifting, which showed a similar result as the former study.

Burgess-limerick et al. (1995) studied the inter-joint coordination with varied load weights in the symmetric lifting task. Five load weights from 2.5 to 10.5 kg with 2 kg increments were lifted by each participant. Participants were instructed to use a self-selected lifting style, and they adopted a lifting strategy between stoop and full-squat. The coordination was quantified by the relative phase angles between the adjacent joint angles. It was observed that knee extension led hip extension, which in turn led the extension of lumbar vertebral joints. As the load weight increased, the relative phase between the adjacent joints increased significantly, which was in agreement with the results on joint coordination from Scholz (1993) and Scholz et al. (1995). With an increased phase lag of the trunk extension, the erector spinae muscle was longer and shortening more slowly in the early phase of lifting when the load acceleration was high. This pattern in the erector spinae muscle enhanced back extensor strength, which would reduce risks of LBP when lifting loads of higher weight.

Catena et al. (2010) examined the effects of load weight on the postural stability during a 180° lateral load transferring task. Participants transferred a loaded (5% of body weight) and an unloaded box. The projection of the whole-body centre of mass (COM) onto the base of support (BOS) and the muscle activities of lower extremity muscles were measured to assess the postural stability. The increased load weight resulted in significantly decreases in the horizontal distance ($\geq 1\text{cm}$) between the COM and the BOS edges. Besides, increased COM movement (up to 14%) and higher muscle activities on both sides of the shank were also found with increased load weight. Therefore, the increased load weight induced higher risks of balance loss.

Yoon et al. (2012) studied the effects of load weight and lifting speed on the trunk and upper extremity muscle recruitment. Participants lifted two load weights which were the 'light' load (larger value between 25% of maximum lifting capacity and 3 lbs) and the 'heavy' load (60% of maximum lifting capacity) at two lifting speeds (self-selected and maximal). The results

showed that the increased load weight induced significant higher muscle recruitments at the trunk and upper extremities throughout the whole lifting period without changing the temporal pattern. The increased lifting speed, however, significantly changed the temporal pattern in most of the muscles. In particular, the higher lifting speed increased the muscle recruitment at the initial phase of lifting, whereas decreased the muscle recruitment at the terminal phase of lifting.

2.2.1.5 Load Knowledge

It is often suggested that lifting loads with unknown weight could be a risk factor of LBP. Underestimating and overestimating load weight may result in increases of muscle activities and loss of balance. Commisaris and Toussaint (1997) examined the effects of overestimating load weight on low back load and postural balance during lifting. In a series of lifting tasks, the lifted load weight was unexpectedly changed from 16kg to 6kg. The low back load was quantified with the lumbosacral joint moment, and the postural balance was studied by the position of the whole-body centre of gravity (COG) relative to the base of support, the linear momentum of the COG and the angular momentum of the whole body. In spite of 10 kg decreases in the load weight, the result showed no significant difference in the low back moment between lifting 6 kg and 16 kg during the first 150 ms after the load was lifted. Moreover, overestimating the load weight led to a significant overshoot in the linear and angular momenta, which caused higher risks of balance loss. Therefore, overestimation of the load weight would lead to higher low back load and increased risks of losing balance during lifting.

Burg et al. (2000) examined the effects of lifting an unexpected heavy load on low back load and balance control. Participants were instructed to lift a box as quickly as possible. The weight of box was unexpectedly increased by 5 or 10 kg in some trials during a series of

lifting tasks. Trunk kinematics, the linear and angular momenta of the whole-body centre of mass (COM) and the L5/S1 joint torque were measured. The results showed that lifting an unexpected heavy load led to decreases in the peak L5/S1 joint torque (4.5% - 7.2%) compared to lifting the same load weight with correct expectation. Risks of balance loss appeared to be increased since the angular momentum was significantly lower in the unexpected load weight condition than that in the correct expectation condition.

In another study, Burg and van Dieën (2001) also investigated whether underestimating the load weight would increase low back load during lifting. Participants were instructed to lift boxes weighting 1.6 or 6.6 kg at a self-selected speed. In some trials, the weight of the box increased by 10 kg unexpectedly. The results showed that underestimation of load weight did not increase the L5/S1 torque and compression force.

Watanabe et al. (2013) examined the effects of lifting a heavy weight than expected on trunk muscle activities. Three task scenarios were included: lifting an expected 1-kg load, lifting an unexpected 4-kg load (expected to be 1 kg) and lifting an expected 4-kg load. The results showed significant effects of underestimating the load weight on the temporal patterns of the muscle activities. Specifically, when participants lifted an expected load weight, trunk muscle activities increased immediately after the onset of lifting; when participants were unaware of the load weight, the increases in muscle activities were significantly delayed. Therefore, lifting loads heavier than expected resulted in a delayed response in trunk muscles which might be associated with increased risks of LBP.

Kotowski et al. (2007) studied the effects of load weight sequence and knowledge on trunk kinematics during repetitive lifting. Three load weights (4.54, 9.07 and 13.61 kg) were used. Participants performed three different lifting tasks: random unknown (RU), random known (RK), and same weight (SW). In the RU task, the sequence of the load weights was

randomized, and no weight information was informed to participants before lifting. For the RK task, the lifted load weights were in a random order, but participants were informed of the load weight before lifting. During the SW task, a series of equal numbers of same-weight loads was lifted in sequence (e.g. all 4.54 kg, all 13.61 kg, then all 9.07 kg), and the load weights were not informed to participants. By comparing the RU and RK tasks, the effects of load weight knowledge can be determined; by comparing the RU and SW tasks, the effects of load weight sequence can be examined. The results showed significant effects of load weight knowledge and sequence on trunk kinematics. However, the differences were relatively low in magnitude.

2.2.1.6 Fatigue

Muscle fatigue is common in manual tasks. It decreases the strength and working capability of workers and may consequently lead to increased risks of injuries. In lifting tasks, fatigue has also been reported to have significant influence on lifting kinematics and spinal loadings. Some studies on the effects of various fatigue protocols are summarized in this section.

Sparto et al. (1997) studied the changes of the lifting kinematics and postural stability caused by fatigue in repetitive lifting. In the study, participants lifted repetitively a load of 25% of their maximum lifting capacity at a maximum rate in their self-selected style. The results showed significant decreases in knee (32.6%) and hip (10.3%) ranges of motion and increases in spine peak flexion (13.8%) after fatigue. Fatigue also caused greater hip and lumbar spine extension at the early lifting phase. The anterior-posterior excursions of the trunk centre of mass and the centre of pressure increased significantly with fatigue by up to 40.3% and 50%, respectively, indicating that the postural stability declined as fatigue was induced.

In another study, Bonato et al. (2002) explored how fatigue in repetitive lifting affected whole-body kinematics and muscle activities. Participants lifted the load of 15% of their body mass at a rate of 12 lifts/min for 5 minutes. A decrease in the back muscle activities was observed after fatigue. The results also showed significant increases in the lifted load acceleration after fatigue, which induced higher torque at L4-L5 the joint. An increase of knee and elbow range of motion was also found and suggested to be a biomechanical adaption to back muscle fatigue.

Dolan and Adams (1998) investigated the effects of fatigue induced by repetitive lifting on the spinal loads. In the experiment, participants lifted and lowered a 10 kg load from floor to waist height 100 times. Fatigue resulted in 36% increases in the peak bending moment on the lumbar spine but 11% decreases in the peak spinal compression force at the lumbar spine.

Van Dieën et al. (1998) conducted a study on the effects of fatigue induced by repetitive lifting on whole-body kinematics. Participants lifted a load weighted at 10% of body mass for about 40 minutes. The trunk extension velocity decreased ($p = 0.003$) and the hip extension velocity increased ($p = 0.042$ for the left hip and $p = 0.043$ for the right hip) significantly after fatigue. These changes caused an increased phase lag between the hip and trunk extension. Also, fatigue caused changes in the initial lifting posture: significantly larger leg extension (e.g., $p = 0.001$ for the right ankle and $p = 0.01$ for the right knee) and trunk flexion ($p < 0.001$) were observed after fatigue. The authors argued that the change in the initial posture was an adaption to the fatigue and was interpreted as a function of human body to hinder further fatigue development.

Mehta et al. (2014) examined the effects of fatigue on physiological and biomechanical responses during a repetitive asymmetric lifting task. In the experiment, participants repeatedly lifted loads weighted 15% of their maximum lifting strength at the rate of 10

lifts/min. The load origin was located at the ankle height in the sagittal plane, and the load destination was set at the waist level at the left side of the participants. The results showed significant increases in subjective workload measured by the Borg CR-10 scale, and significant decreases in spine muscle oxygenation levels over time. Fatigue also induced significantly larger trunk flexion angle, sagittal extension and lateral bending velocities. All these fatigue-induced biomechanical changes are associated with increased risks of LBP.

In addition to the whole-body fatigue induced by repetitive lifting, the effects of localized fatigue induced by specific protocols have also been studied. It has been reported that the relative strength of muscle groups would influence the lifting techniques (Zhang and Buhr 2002). Since localized fatigue might change the relative strength of different muscle groups, it could make a difference in the lifting styles as well. Trafimow et al. (1997) studied the effects of quadriceps fatigue on lifting styles. The trunk angular velocity increased, and the time integral of knee moment over the lifting period and the hip angles decreased significantly after quadriceps fatigue, which indicated a change in lifting styles from squat to stoop.

In another study on quadriceps fatigue, Sasaki et al. (2008) investigated the changes of joint angles and muscle activities in different levels of fatigue. Participants performed maximum isometric extension of both legs to induce the quadriceps femoris fatigue. Three levels of fatigue were induced, including 0% (no fatigue), 25% (median level) and 50% (high level) fatigue. Participants were instructed to perform lifting tasks in self-selected manners. At 25% fatigue, the knees were extended more and the trunk was leaned further forward compared with 0% fatigue. From 25% to 50% fatigue, the joint angles did not change significantly. Decreased lumbar muscle activity was observed at 25% fatigue compared with 0% fatigue, while the lumbar muscle activities were higher at 50% fatigue than that at 0% fatigue. It was argued that at the lower level of fatigue, human body was able to alter the lifting kinematics

to maintain the low back load at a low or median level, but when the fatigue level became higher, the compensatory changes could not be made and eventually the low back loads were increased.

Chen (2000) investigated the effects of localized arm fatigue on lifting kinematics and low back loads. The localized arm fatigue was induced by a static posture protocol in which the participants hold a weight with their arms flexed 90° as long as possible. Participants performed a symmetric lifting task in free style before and after the fatigue protocol. The results showed the lifting style was closer to stoop after fatigue. The participants tended to adopt an accelerated technique in which the peak load accelerations were increased after fatigue. This fatigue-induced change in lifting kinematics resulted in significantly higher compressive forces at the L5/S1 disc and was more likely to induce low back injuries.

Gorelick et al. (2003) examined the effects of muscle fatigue induced by two different fatigue protocols on the coordination of trunk and thigh muscles during stoop lifting. The two fatigue protocols were: 1) a non-specific whole-body fatigue (rowing fatigue protocol) and 2) a specific fatigue on trunk extensor muscles (back extension fatigue protocol). The results showed no significant effects of the rowing fatigue protocol on both trunk and thigh muscle activities. The back extension fatigue protocol produced significant changes on the timing of the trunk muscle activation during lifting. Specifically, the trunk muscle activations started earlier and had longer durations after the back extension fatigue protocol.

2.2.1.7 One- and Two-handed Lifting

Lifting using one hand is very common in industry, and may cause asymmetric effects on motion patterns and the spinal loads. Allread et al. (1996) studied trunk kinematic differences between unsupported one-hand and two hand lifting. Participants lifted loads of varied weights (3.4, 6.8 and 10.2 kg) from different asymmetric starting positions (sagittal

symmetric and 45, 90 and 135 degree to the right of the mid-sagittal plane) with one hand or two hands. It was found that one-handed lifting resulted in significantly larger trunk motion ranges in the lateral (increased by 108.5%) and transverse (increased by 25.1%) planes and greater maximum trunk flexion (increased by 12.6%) in the sagittal plane. Also, the maximum and average trunk velocities in the lateral plane (with increases of 49.0% and 25.1%, respectively) and the average trunk velocity in the transverse plane (increased by 13.8%) were significantly larger in one-hand lifting. As suggested by Marras et al. (1993), the large maximum sagittal flexion, maximum lateral velocity and average twisting velocity in the lumbar spine motion would result in high risk of LBP. Therefore, unsupported one-handed lifting should be considered as a potential risk factor for low back injuries.

Kingma and van Dieën (2004) studied the trunk kinematics and the internal loads at the L5/S1 joint in three lifting styles: two-handed lifting and two types of one-handed lifting (with or without supporting the upper body with the free hand). Participants were instructed to lift loads over an obstacle in the sagittal plane. The results showed that the two-hand lifting resulted in significantly higher peak trunk flexion, but lower peak trunk lateral bending and twisting compared to both one-hand lifting tasks. The peak total net moments, compression and shear forces at the L5/S1 joint were significantly lower in the supported one-hand lifting compared with the two-hand lifting. These results suggested that supported one-hand lifting reduced low back loads but increased asymmetry in lumbar motions.

2.2.1.8 Foot Stance Conditions

Foot stance conditions affect the base of support during lifting, and might further influence postural control and lifting biomechanics. Kollmitzer et al. (2002) examined the effects of stance condition on postural control during lifting tasks. In the experiment, participants lifted and lowered a load of 4.6 kg between a desk and a shelf 10 times during 1 minute. Two

stance conditions, including parallel and step conditions, were involved. Lifting kinematics and ground reaction forces were examined. The parallel stance resulted in significantly higher ground posterior shear forces than did the step stance ($p \leq 0.001$), therefore posed increased challenge in postural control in the sagittal plane. The step stance increased lateral body movements ($p \leq 0.02$) and lateral ground shear forces ($p = 0.007$) compared with the parallel stance. Therefore, both stances have positive and negative effects on postural control.

Brown and Reiser II (2012) studied the effects of straddle stance on the limb load distribution during lifting. Participants lifted a box (17 kg for men and 12 kg for women) from the floor to the standing hip height. Three straddle stances were examined: dominant foot forward (DFF), non-dominant foot forward (NDFF) and a symmetric stance (SYM). The limb load was defined as the percentage of the total time-average ground reaction force under each limb. The results showed significant effects of straddle stances on the load distribution between two limbs. Specifically, the load differences between the dominant and non-dominant limbs ($D\% - ND\%$) in DFF, NDFF and SYM stances were - 6.9%, 2.1% and -1.3%, respectively. After adjusting load asymmetry between two limbs based on the load distribution in the SYM stance, the trail limb generated significantly higher loads than did the lead limb in all three stances, although such limb load asymmetry was mitigated in the NDFF stance compared with the DFF stance. Therefore, the NDFF stance should be used in order to minimize the difference between the two limb loads.

Sorensen et al. (2011) investigated the effects of stance width on trunk kinematics and muscle activities during lifting. Participants lifted and lowered a load of 10 kg in the sagittal plane under three stance width conditions: narrow (feet together), moderate (the same as shoulder width) and wide (150% of the shoulder width). The results showed that the increased stance width significantly reduced peak trunk sagittal flexion angles and angular accelerations

($p < 0.001$) which might lead to decreased low back loads. No significant effects of stance width were found on the trunk and lower extremity muscle activities.

2.2.1.9 Use of Assistant Tools

Some researchers investigated the influence of the use of assistant tools and devices on the lifting motions and spinal loads. Marras et al. (2000) explored the effect of an elastic lumbar back support on the spinal loading and lifting kinematics. In this study, participants performed lifting tasks (including both symmetric and asymmetric tasks) with different load weights (13.6 kg and 22.7 kg) in their self-selected styles, and allowed to move their feet during the task. Significant decreases in the maximum trunk sagittal flexion angle (36.5 to 32.7°) and extension velocity (47.2 to 40.2°/s) were observed when participants wore the lumbar back support. However, no significant difference in the net moments and spinal forces at the L5/S1 joint was observed between wearing-a-lumbar-back-support and no-back-support conditions. Therefore, given the unchanged spinal loads, wearing a lumbar support provided no protective effect when lifters were allowed to move their feet in lifts.

Queen's University (Kingston, Canada) developed a personal lift-assist device (PLAD) which generates external forces using stored elastic energy (Abdoli-E et al. 2006). Energy is stored when the elastic elements are compressed during trunk flexion, and the stored energy is released when the trunk is extended and the load is lifted up. The effects of using this device during lifting tasks have been examined extensively. It was found that this device is effective in increasing dynamic stability of the spine, and reducing low back loadings, back muscle activity levels and muscle fatigue during various lifting task scenarios (Abdoli-E and Stevenson 2008; Frost et al. 2009; Godwin et al. 2009; Graham et al. 2011; Lotz et al. 2009; Whitfield et al. 2014). The PLAD also induced significantly decreased lumbar and thoracic flexion and significantly increased hip and ankle flexion (Sadler et al. 2011). All these effects

caused by wearing the PLAD indicated that using the device is effective to reduce risks of LBP during lifting.

Elford et al. (2000) investigated whether the use of lifting slings reduced the risk of injuries in patient handling tasks. The participants adopted three lifting techniques (no slings, one and two slings) in the patient handling task. The lumbar kinematics, lifter preference and subjective ratings of body part stress were used as the measurements to assess the risk of injuries for three techniques. The kinematic results showed that the angular displacements, velocities and accelerations of the lumbar spine were significantly greater ($p < 0.05$) in the sagittal, transverse and lateral planes for no-sling lifts compared to the other two techniques. The stress ratings of all body parts were significantly higher in the no-sling lifts, especially for the low back and abdomen parts. All participants showed their preference for lifts with slings. Therefore, using slings in patient handling techniques were recommended.

2.2.1.10 Ground Surface Characteristics

In some work environments (e.g. in agriculture and construction workplaces), workers need to perform lifting tasks on the ground surface that is slippery or not perfectly flat. These ground surface characteristics may have impacts on lifting kinematics and spinal loads, and induce higher risks of low back injuries and balance loss.

Shin and Mirka (2004) investigated the changes of the spinal loads in lifting tasks on the ground surfaces with different slope angles in the sagittal plane. Five slope angles were included: two inclined slope angles (facing an upward slope: 20° and 10°), two declined slope angles (facing a downward slope: -20° and -10°), and a flat surface (0°). Participants were instructed to use squat, stoop and freestyle lifting styles to symmetrically lift a 10 kg box. The results showed significant effects of slope angles on the peak net moment at the L5/S1 joint in the squat and freestyle lifting, but no effects on the moment in the stoop lifting. In the

freestyle lifting, the peak net moment was significantly higher for the inclined surfaces as compared with the flat surfaces ($\geq 6.8\%$) or declined surfaces ($\geq 10.0\%$). While in the squat lifting, the minimum net moment occurred when lifting on the flat ground surface.

Jiang et al. (2005) studied biomechanical responses in the dynamic lifting on the laterally slanted ground surface. The participants performed the lifting tasks on a platform that was laterally tilted at 0° , 10° , 20° and 30° from horizontal. With increased slope angles, the results revealed a consistent reduction in the peak dynamic L5/S1 moment (decreased by 9% from horizontal to 30°), but significantly increased peak lateral forces at the L5/S1 joint (increased by 111%) which indicated a higher risks of instability.

Catena et al. (2011) investigated balance control strategies to negotiate a slippery surface during load transferring. In the experiment, participants performed a 180° lateral load transfer task over high- and low- friction surfaces ($\mu = 0.86$ and $\mu = 0.16$ respectively). Participants were not allowed to move their feet, but could pivot their feet during lifting. Balance measures, body kinematics and lower extremity muscle activities were examined. The higher slippery surface resulted in significant increases in the travelling distance of the whole-body centre of mass (COM), and significant decreases in the horizontal distance between the COM and the edges of base of support. To enhance body balance on the slippery surface, participants reduced the stance width by 8cm and rotated external foot by 15° towards the load. No significant changes on the lower extremity muscular control were observed to adapt to a slippery surface.

2.2.1.11 Other Work-related Factors

Other work-related factors including load handle characteristics, sudden perturbations, lifting on a ship, etc., have also been examined in biomechanical analysis on manual lifting.

The load handles serve as the interface between the lifters and the lifted boxes in lifting tasks, and may have effects on the kinematics and lifting capacity. Wang et al. (2000) studied whether the handle angle affected the lifting capacity and wrist postures. Six handle angles equally spaced from 0° to 75° were studied in this experiment. No significant effect of handle angle on maximal acceptable weight of lifting (MAWL) was observed, but wrist postures were considerably affected by the handle angle. Two extreme handle design, 0° and 75° handle angle, produced greatest ulnar and radial deviation, respectively. The medium handle angles, such as 30° and 45° , were recommended since they produced the most neutral wrist postures. Haddad and Mirka (2010) investigated how the low back injury risk was influenced by the hand-hold locations on the load in lifting tasks. Four hand-hold locations (two symmetric and two asymmetric) were adopted, and the trunk kinematics were used as the input of a lumbar motion monitor (LMM) risk assessment model (Marras et al. 1993) to calculate the risk index for LBP. The results showed that holding the box symmetrically at the top produced the least risk of low back injuries (18% decreases compared with the asymmetric condition).

Chow et al. (2005) studied the muscular and postural reactions to the sudden release of the load during lifting. Participants performed symmetric squat and stoop lifts with loads from 20 to 80N, and were perturbed with sudden release of the load in some random trials in which the actual load magnitude was decreased. The sudden release triggered the same sequence of muscular activity in both styles. The posture disturbance appeared to be greater for squat lifting than stoop lifting at the higher loads of 60 and 80N, since the center of pressure moved closer to the posterior limit of stability (the line joining the heels) in the squat lifting. Therefore, squat lifting strategy may induce higher risks of balance loss when a load weighted over 50 N was suddenly released.

The risk of injury might increase when workers are performing lifting tasks on a ship. An example was given by Faber et al. (2008) who investigated the spinal loadings when participants performed repetitive lifting tasks in a sailing ship. Participants lifted an 18 kg box, and the whole body kinematics, ground reaction forces, EMG measurements and the ship motion (the vertical and horizontal accelerations of the ship) were recorded. The effects of the ship motion (the accelerations) on the spinal loads were studied. The results showed significantly increased net moments and compressive forces at the L5/S1 joint with the increased vertical ship acceleration.

Ning and Mirka (2010) studied the effects of ground motions on lifting biomechanics. Participants lifted loads of two weights (5 and 10 kg) on a boat motion simulator which simulated the deck motion of a small fishing boat. Five ground motion conditions defined by different ground angular displacements and vertical accelerations (A: $+6^\circ$, -0.54 m/s^2 ; B: $+3^\circ$, -0.27 m/s^2 ; C: 0° , 0 m/s^2 ; D: -3° , 0.27 m/s^2 ; E: -6° , 0.54 m/s^2) were examined. Trunk kinematics and trunk muscle activities were studied. The results showed significantly higher peak trunk sagittal angular accelerations in condition A than those in conditions C, D and E ($698^\circ/\text{s}^2$ vs. $612 - 617^\circ/\text{s}^2$). The external oblique muscles showed significantly higher activity levels in the moving conditions (A and E) compared to the stationary condition (C), which indicated this muscle group played an important role in maintaining torso stability during the moving conditions.

Jin and Mirka (2011) examined the effect of a lower extremity kinematic constraint on lifting biomechanics. In the experiment, a thigh-level railing was set in front of participants to impose constraints on knee and ankle joint movements. Participants negotiated with the barrier using two strategies (leaning and no leaning on the railing) when lifting loads from three height levels (85, 70 and 55 cm from the ground). Trunk kinematics, muscle activities and ground reaction forces were examined. The results showed significant decreases in

muscle activities at the erector spinae (36% vs. 26% MVC) and latissimus dorsi (14% vs. 8% MVC) when using the leaning strategy compared with the no-leaning strategy. The leaning strategy also resulted in significantly lower peak trunk sagittal accelerations, but significantly higher required coefficient of friction (0.05 vs. 0.36). Therefore, the leaning strategy may reduce low back load, but increase risks of slips compared with the no-leaning strategy.

2.2.2 Personal factors

Due to the individual differences (e.g. age and gender differences), the same manual task condition can result in different impacts on body kinematics and kinetics among varied populations. Therefore, personal factors (i.e. individual differences) should be taken into account when considering the ergonomic task design so that the design is able to accommodate different populations. The most common personal factors under investigation include gender, work experience, muscle strength and age. The effects of these factors on lifting biomechanics are summarized below.

2.2.2.1 Gender

The females and males differ greatly in their body sizes, muscular strengths and working capacities, which leads to considerable distinctions in the performance during lifting, such as lifting techniques (Lindbeck and Kjellberg 2001; Stevenson et al. 1996), spinal loading (Marras et al. 2002, 2003) and metabolic cost (Kell and Bhambhani 2003). Therefore, the kinematic and kinetic differences between genders in lifting tasks should be examined and taken into account in the workplace and task design, so that the design will be suitable for both genders.

Lindbeck and Kjellberg (2001) investigated the gender effects on lifting kinematics. Participants lifted a load from floor to their navel height with both squat and stoop lifting techniques in the sagittal plane. The load weight was 12.8 kg for the male participants and 8.7

kg for the female. The results showed that the male participants had significantly larger trunk ranges ($p=0.019$) of motions in both lifting styles and the female tended to have larger knee motion ranges ($p=0.0083$) in squat lifts. The knee-hip coordination in squat lifts was examined and quantified using the relative phase angle between the two joints. The relative phase angle was significantly smaller in the females than that in the males, which indicated the females showed more synchronized motion pattern than did the males.

In two other studies (Albert et al. 2008; Sadler et al. 2013), however, no significant gender effects on lifting kinematics were observed. In Albert et al.'s study (2008), participants lifted loads weighed 20% of their personal maximum lifting capacity from the floor to the waist height in a freestyle manner. The results showed no gender-related significant difference in the coordination patterns between the knee, trunk and shoulder joints in terms of their relative phase angles. Sadler et al. (2013) also investigated the gender effects on the lifting techniques. Participants lifted loads of 0% and 10% of their maximum back strength to 50% of their individual heights in the sagittal plane. The joint angular trajectories were analysed using a principle component analysis. The first 30 principle components (PCs) which retained 90% of the variance were extracted. No significant effect of gender on these PCs was found. Therefore, the authors suggested that both the female and male participants adopted the same lifting techniques. The results on gender effects reported by Albert et al. (2008) and Sadler et al. (2013) appears to contradict with those from Lindbeck and Kjellberg (2001). This discrepancy might be due to the differences in the load weight selection protocols among three studies. In Albert et al. (2008) and Sadler et al. (2013), load weights were normalized for each participant according to their individual lifting capacity; while in Lindbeck and Kjellberg (2001), all male and female participants were assigned with loads of 12.8 kg and 8.7 kg, respectively.

Rae et al. (2013) examined biomechanical differences between genders during a fatiguing asymmetric lifting task. Participants lifted loads weighted 10% of participants' maximum lifting capacity for 75 minutes at a rate of 6 lifts/min. The load origin was located at the ground right ahead of participants, and the load destination was set at 60° to the right of the sagittal plane and at the ASIS height level. Lifting kinematics changed differently in the females and males with fatigue. The females showed significantly increased ranges of motion in right shoulder in/external rotation and ab/adduction after fatigue. For the females, fatigue also induced significantly increased peak angular velocities in the right elbow and left shoulder flexion/extension, as well as in the right shoulder in/external rotation. The males had fewer kinematic changes caused by fatigue. Specifically, the motion ranges in the trunk flexion/extension and lateral bending were increased after fatigue in the males.

Marras et al. (2003) investigated the gender effects on the spinal loading during realistic lifting tasks. Both male and female participants conducted symmetric and asymmetric lifting with three levels of load weights (6.8, 13.6 and 22.7 kg). It was found that the males generated significantly larger compression force and anterior-posterior (A-P) shear force at low back part than did the females (about 640 N and 300 N, respectively). Such gender-related differences in the spinal loadings appeared to be a result of the differences in lifting kinematics, muscle activities and body weights between two genders. Specifically, the females showed significantly larger hip flexion, while the males adopted significantly greater lumbar flexion. As for the muscle activities, the females showed a significantly higher level of activities of extensor muscles compared with the males. Although the males suffered greater spine loads than the females, the females were still in a higher risk of LBP due to their relatively lower tissue tolerance on the spinal loads.

Nimbarte (2014) examined the effects of gender on neck muscle activities during lifting exertions. Four lifting heights (knuckle, elbow, shoulder and overhead) and three load

weights (25%, 50% and 75% of participants' maximum strength) were involved in the experiment. The activities of cervical trapezius and sternocleidomastoid muscles were examined. The results showed significantly higher sternocleidomastoid muscle activities and significantly lower cervical trapezius muscle activities in females compared with males.

2.2.2.2 Work Experience

Differences in lifting strategies between novice and experienced workers have been found in previous research (Authier et al. 1995, 1996). They found that experienced lifting workers tended to adopt strategies that were safer and more efficient compared with their novice counterparts. Therefore, comparing the lifting kinematics between experienced and novice handlers may help identify the optimal lifting strategies.

Gagnon et al. (1996) investigated the differences of lifting strategies between experienced workers and novices, and determined the effects of work experience on spine loads. Both the experienced workers and novices performed asymmetric lifting tasks with permission to move their feet. The lifted loads (12 and 22 kg) were transferred from four different initial positions to a platform on a trolley. The results showed no significant differences in trunk kinematics or net moments at the L5/S1 joints between two groups. However, significantly reduced knee flexion (31° vs. 53°) and excursion (20° vs. 38°), and decreased asymmetry of movements between two knees were observed in lifting trials of the experienced workers compared with those of novice ones. The experienced workers' strategy appeared to be energy efficient and safer because the reduced knee movements resulted in decreased energy expenditure and increased knee stability.

Plamondon et al. (2010) compared the lifting kinematics and low back moments of the experience and novice workers in an asymmetric lifting task to verify whether the lifting strategy of the experience workers was safer and more efficient than that of the novices. In

the experiment, two load weights (15 and 23 kg) were adopted, and participants were free to choose handling techniques and velocities. The results showed no significant differences in the net moments at the L5/S1 joint between the experienced and novice workers. However, the experienced workers bent their lumbar region less (experts 54° vs. novices 66°), but bent their knees more (experts 72° vs. novices 53°), which brought them closer to the load. Besides, significantly higher lumbar extension velocity was also found in the experienced workers than that in the novices.

Plamondon et al. (2014) studied the differences in lifting motions and back loading between experienced workers and novices during a repetitive box transferring task. In the experiment, participants were instructed to transfer multiple 15-kg boxes from one pallet to another in two different paces (self-determined and 9 lifts/min). As found by Plamondon et al. (2010), the experienced workers showed significantly reduced lumbar spine flexion compared with novices (10° less). No significant differences in the low back moments between experienced and novice groups were found. Therefore, the lifting posture was the key biomechanical difference between experience workers and novices.

Granata et al. (1999) examined the effects of work experience on trunk kinematics and spinal loads during repeated lifting tasks. Two load weights (13.6 and 27.3 kg), two levels of task asymmetry (sagittally symmetry and 60° to the right of the sagittal plane), and two lifting velocities (self-selected and faster than self-selected) were involved in lifting tasks. The results showed that the experienced workers adopted significantly smaller trunk motion ranges (7.9° less) and peak trunk angular velocities (7.8°/s less) in the sagittal plane than did the novices. The experienced workers also showed significantly great spinal loads (trunk moments in the sagittal and transverse planes, spinal compression and shear forces) than the novices (20.3% - 49.1%).

Marras et al. (2006) investigated the spine loading as a function of lifting frequency, exposure duration and work experience. Participants performed repetitive asymmetric lifting for 8 hours. Spinal compression and shear forces were examined as spine loads. The results showed significantly greater spinal compression forces ($p=0.0043$) in the novices compared with the experienced workers. It was also found that the spinal loading increased after the first 2h of lifting exposures regardless of the lifting frequency.

Lee and Nussbaum (2012) also studied the differences in lifting biomechanics between experienced and novice workers during repetitive lifting and lowering tasks. Participants performed both symmetric and asymmetric lifts and lowers with a box weighted 10% of individual body mass. The results showed significantly higher peak torso kinematic and kinetic measures in the experienced workers compared with the novices, which suggested a relatively higher risk of LBP in the experienced group. Lee and Nussbaum (2013) further compared the dynamic balance and torso movement stability between the two groups using peak linear/angular momenta and largest Lyapunov exponent (LLE) of torso flexion and extension. The results showed that horizontal linear momenta, all angular momenta and LLEs were significantly lower in the experienced workers. This suggested that experienced workers had better body balance than the novices. Therefore, it was suggested that experienced workers might sacrifice torso kinematics/kinetics to enhance dynamic balance during repetitive lifting.

The above studies showed contradictory results on lifting biomechanics when comparing experienced and novice workers. For instance, Gagnon et al. (1996) found reduced knee flexion in experienced workers, yet Plamondon et al. (2010) observed that experienced workers bend their knee more than did the novices. Granata et al. (1999) reported significantly higher spinal loading in experience lifters, whereas other studies (Gagnon et al. 1996; Marras et al. 2006; Plamondon et al. 2010) showed comparable or even lower spinal

loads in the experienced lifters than the novices. The mixed results might be caused by the different task scenarios in these studies. In fact, Lee and Nussbaum (2012) reported that experienced workers modified their lifting strategies with varied task conditions, whereas the lifting patterns of the novices were more consistent.

Delisle et al. (1999) studied the different footstep strategies adopted during lifting tasks between experienced workers and novices. In the experiment, participant lifted and lowered a load (weighted 12 kg) from a low cart (0.10 m from the ground) to a shelf (1.20 m), and the cart and shelf formed a 90° angle. Two footstep strategies (called large-step and backward-step) were found from experienced workers and another two (called oblique-step and cross-step) were found from novice workers based on the observation in the experiment. The results showed the experienced workers' strategies were effective to either reduce the path of the centre of gravity (COG) or reduce the asymmetry of the body posture at deposit. One of the benefits of reducing the path of the COG is reducing the required effort and avoiding fatigue.

Gagnon (2005) summarized studies on comparing the different box maneuverer strategies between experienced and novice workers. It was found that the experienced workers tended to tilt the box before lifted it. In symmetric lifts, the experienced workers adopted parallel handgrips (both hands on the lateral edges under the box) and used a backward tilt (towards the workers). In asymmetric lifts, they preferred diagonal handgrips. Compared with no-tilt strategies adopted by the novice workers, the experienced workers' strategies reduced the efforts since the knee flexion and load transfer duration were both reduced.

Plamondon et al. (2012) compared the effect sizes of three factors (i.e. work experience, lifting height and load weight) during manual handling tasks. Participants transferred a box from a conveyor to a hand trolley during the experiment. Two load weights (15 and 23 kg)

and two lifting heights (ground level and 32 cm from the ground) were examined in the experiment. The results showed that all three factors had significant effects on lifting postures and lumbar external moments. Specifically, the experienced workers showed significantly smaller lumbar flexion ($p < 0.01$) and larger left knee flexion ($p \leq 0.05$) compared with the novices. Lifting height and load weight had larger effect sizes than work experience on lumbar external moments. Therefore, in order to reduce low back load, ergonomic interventions should be first focused on optimizing lifting height and load weight.

2.2.2.3 Muscle Strength

Muscle strength is an important determinant of load carrying capacity thus should be considered when designing the lifting task (Haisman 1988). For instance, light weights should be assigned to the workers with weaker muscle strength so as to reduce the risks of injuries. In this section, several studies investigating the effects of muscle strength on lifting biomechanics are summarized.

Li and Zhang (2009) investigated whether the relative strength between the back and knee joints would influence the lifting strategies. In the study, participants first conducted isokinetic strength tests so that they were divided into two groups: a back-stronger-than-knee group (BSTK) and a knee-stronger-than-back (KSTB) group. The participants were then instructed to perform symmetric lifting using their own preferable lifting strategies and velocities. A postural index proposed by Burgess-Limerick and Abernethy (1997b) was adopted to quantify the lifting strategies. The result showed significant effects of the relative strength on choosing the lifting strategy: subjects within BSTK group were more likely to use back lift while subjects in KSTB group tended to adopt leg lift. Therefore, muscle strength is a determining factor of lifting strategy.

Zhang and Buhr (2002) studied whether the strength of leg and back muscles affected the lifting strategies. The participants performed lifting tasks in three conditions: 1) the unconstrained condition without instructions on lifting strategies; 2) the gastrocnemius weakness condition in which the contraction level of gastrocnemius was constrained at 45% of its maximum strength using an EMG-based biofeedback technique and 3) the rectus femoris weakness condition in which the contraction level of rectus femoris was controlled in the same way as in the second condition. The results showed that both of the simulated muscle weakness conditions significantly influenced lifting strategies. The participants who used a leg-lift strategy in the unconstrained condition were observed to use a back lift in the other two conditions in which the leg strength was weakened.

2.2.2.4 Age

LBP is more prevalent among older adults (Holmstrom and Engholm 2003; Peek-Asa et al. 2004; Heiden et al. 2013). Given that the workforce is aging worldwide, there is a need to study age-related differences in lifting biomechanics. Shin et al. (2006) examined age-related changes in trunk kinematics and ground reaction forces during lifting. Participants lifted a load weighted 5 kg from two origins of different asymmetric conditions (sagittally symmetric and 60° to the right of the sagittal plane) on the ground to three destination heights (knee, waist and shoulder levels) in the sagittal plane. The results showed that older adults had significantly lower peak twisting velocities (40% less) and accelerations (30% less) in the transverse plane. No significant age-related differences were observed in ground reaction forces and other trunk kinematics measures.

Burgess et al. (2009) investigated the age-related differences in multi-segment trunk motions during a symmetric lifting task. Participants lifted a load of 5 kg from bench to shelf under two stances (step and parallel). Older adults showed significantly higher linear pelvic and

trunk displacements and smaller ranges of pelvic and lower thorax angular displacements compared with younger ones. The reduced angular displacements in pelvis and lower thorax might result in more forward trunk postures, and therefore increased risks of falling in older adults.

2.2.2.5 Other Personal Factors

Some other personal factors, such as obesity, having LBP, etc., also significantly change the lifting biomechanics. Xu et al. (2008) examined the effects of obesity on trunk kinematics and ground reaction forces during lifting. Two groups of participants were recruited, including the normal weight group (body mass index (BMI) $< 25 \text{ kg/m}^2$) and the obese group (BMI $> 30 \text{ kg/m}^2$). Participants lifted loads of different weights (10% and 25% of capacity) from symmetric (in the sagittal plane) and asymmetric (45° to the right of the sagittal plane) origins. The results showed that the obese group had significantly higher trunk velocities (30 - 59% greater) and accelerations (51 – 57% higher) in both of the transverse plane and the sagittal plane. No significant differences between two groups were found in the ground reaction forces which were normalized by individual body weights.

Larivière et al. (2000) compared the differences in trunk coordination between subjects with and without chronic LBP during lifting tasks. During the experiment, participants performed trunk maximal flexion/extension and lateral bending tasks with or without a 12 kg load. During trunk flexion/extension tasks, the patients with LBP showed significantly lower lumbar flexion and significantly higher thoracic flexion compared with the participants without LBP. Therefore, LBP patients increased their thoracic flexion to compensate the decreased lumbar flexion.

Hagins and Lamberg (2011) investigated the differences in breath control between individuals with and without LBP during lifting. Participants conducted lifting tasks with

different load weights (an empty crate and a box of 25% of individual body weights) at a self-selected pace. The amount of volume in lungs was measured using a percentage of each individual's vital capacity (%VC). The results showed that the patients with LBP had significantly larger volume in their lungs (48.2 %VC) compared to the healthy control group (40.9 %VC). This finding suggested that breath control might be a useful intervention in managing LBP for patients.

2.2.3 Limitations in Experiment-based Studies on Lifting Biomechanics

Unlike work-related factors, the effects of some personal factors on the lifting biomechanics have received little attention. In particular, age is one important factor which should be investigated for better occupational lifting task design. Given the global aging workforce and higher prevalence of work-related LBP in older adults (Heiden et al. 2013), the design of the lifting task should accommodate older adults, which calls for research to clarify the age-related changes in lifting biomechanics. Actually, compared with younger adults, older people generally have decreased muscle strength (Doherty 2003; Frontera et al. 2000), reduced joint flexibility (Intolo et al. 2009; Troke et al. 2005) and impaired postural stability (Masani et al. 2007; Sturnieks, George, and Lord 2008). Because of these age-related degenerations, we may expect significant age-related differences in lifting biomechanics (e.g., whole-body kinematics, low back load and postural stability).

However, as reviewed in Section 2.2.2.4, there are only two existing studies (Burgess et al. 2009; Shin et al. 2006) that investigated age effects on lifting biomechanics. One major limitation with Shin et al. (2006) and Burgess et al. (2009) is that they did not report low back load which is a crucial contributor to LBP incidence (Norman et al. 1998). In addition, low back load is significantly influenced by lifting strategies involving whole-body kinematics

(Leskinen et al. 1983). However, age-related differences were only investigated in hip and trunk kinematics by Shin et al. (2006) and Burgess et al. (2009).

Besides, the interaction effects between age and some important work-related factors (e.g. load weight, asymmetric destination) have not been studied. In fact, high load weights have always been considered as a risk factor for LBP (Hoozemans et al. 2008; Plamondon et al. 2012) and loss of balance (Catena et al. 2010), and were found to affect trunk kinematics (Davis and Marras 2000b), inter-joint coordination (Scholz et al. 1995) and muscle recruitment (Yoon et al. 2012) in lifting tasks. Destination height is another work-related factors that influences lifting kinematics (Chen 2002) and low back load (Davis and Marras 2005; Faber et al. 2009a). The effects of these work-related factors on lifting biomechanics and risks of LBP might be influenced by lifters' age because of the age-related physiological changes, such as decreased joint mobility (James and Parker 1989) and muscle strengths (Hurley et al. 1998).

Finally, the only study that has investigated age-related differences in postural stability during lifting was conducted by Shin et al. (2006). A limitation with Shin et al. (2006) is that they did not report measurements of the whole-body center of mass (COM) which provides insight into dynamic postural control mechanisms.

To address these limitations, three experimental studies on lifting biomechanics were conducted and presented in Chapter 4, 5 and 6, respectively. Specifically, the effects of age and its interaction with work-related factors (i.e., load weight and lifting destination height) on whole-body kinematics and low back load were examined in Chapter 4 and 5 for symmetric and asymmetric lifting conditions, respectively. In Chapter 6, age effects on postural stability during lifting were investigated using measurements which were defined based on whole-body COM.

2.3 Human Motion Simulation

Human posture and motion simulation helps human factor/ergonomics specialists to evaluate and improve the product and workplace design in an early phase of production (Chaffin 2005). For instance, in the virtual design environments (e.g. CAD systems), the specialists can use simulated human motions and postures to evaluate human reach capability, vision performance, and other human physical accommodations in industrial workplaces and vehicles, so that the design of these products will be more friendly for their users, and the overall cost of production can be reduced.

Human motion simulation is technically challenging. Because of the redundancy of the human musculoskeletal system, the motion simulation is an ill-posed problem, which means the human motion for a certain task scenario is not unique. For instance, in a hand reaching task, there are infinite hand trajectories and upper limb joint angular displacements connecting the specified initial and final end-effector positions. Extensive researches have been carried out to overcome this ill-posed problem and some of them are reviewed in details in this section.

2.3.1 Optimization-based Motion Simulation

In optimization-based simulation, it is often assumed that the central nervous system (CNS) determines human motions based on certain performance criteria. Therefore, human motion simulation can be formulated as an optimization problem and these criteria are used to define the objective functions in the optimization procedures. However, the mechanisms of how human motions are planned and controlled by the CNS have not been fully revealed and understood. Therefore, there is no consensus on the exact performance criteria adopted by the

CNS. Actually, researchers used various performance criteria to define objective functions in the optimization to generate motion trajectories.

2.3.1.1 Motion Simulation Using Single-objective Optimization

Flash and Hogan (1985) formulated a mathematical model to predict planar arm reaching motions based on a minimum-jerk criterion. The time integral of the square of the jerk (time derivative of the hand acceleration) was minimized in the optimization, which implied the major purpose of the arm coordination was producing the smoothest hand movements. The hand trajectory was represented as two polynomials in time and was predicted when the arm performed unconstrained reaching task in a horizontal plane. The predicted hand trajectory was a straight line, and exhibited a bell-shaped velocity profile. These simulated motion patterns agreed with the actual hand trajectories observed in the experiments. However, this minimum-jerk model only considered the kinematics in hand reaching motions, and the motion dynamics (e.g. muscle forces and inertial effects of arm segments) were not included in the motion simulation.

To consider the motion dynamics, Uno et al. (1989) developed a minimum-torque-change model to predict the hand trajectories in the planar hand reaching task. The objective function was the square sum of the rate of change in muscular torques integrated over the entire movement duration. The hand trajectories generated by this model were different from those in the minimum-jerk model (1985) in some situations. For instance, in hand-point-to-point motions, when the starting point was far away from the human body and the final point was approximately in front of the body, the minimum-torque-change model generated curved trajectories while the minimum-jerk model still generated straight hand paths. The experimental results were closer to the hand trajectories predicted by the minimum-torque-change model than those by the minimum-jerk model, which demonstrated the benefits of

considering the motion dynamics in human motion prediction. However, the computation in the minimum-torque-change model was more intensive because its objective function was defined based on the complex nonlinear dynamics of the musculoskeletal system.

Both Flash and Hogan (1985) and Uno et al. (1989) planned the reaching movements without considering some end-point constraints, such as the direction and velocity of the hand at the target area. Breteler et al. (2001) adjusted the minimum-jerk model and minimum-torque-change model to incorporate these end-point constraints. Right hand motions starting from the left to the right (rightward movements) and right hand motions of the opposite movement direction (leftward movements) were predicted. The approaching angles of leftward and rightward movements were -45° and 45° respectively, and the final speed was 0.8 m/s. The minimum-jerk model produced unimodally-curved trajectories, and the global shape of leftward and rightward movements was similar. However, the trajectories predicted by minimum-torque-change model were S-shaped, and exhibited different shapes between leftward and rightward movements. The hand trajectories measured in the experiment were more similar with the trajectories generated by the minimum-torque-change model than those predicted by the minimum-jerk model.

Mi et al. (2009) developed an optimization-based posture prediction model for human upper body in hand reaching tasks in the three-dimensional space. The upper body was represented as a linkage model including the torso and right arm with multiple degrees of freedom. Three objective functions were involved in the optimization. The first one was minimum joint displacement defined as the difference between the predicted joint angles and those in the most neutral position. The second objective function was minimum effort which was measured as the difference in the joint angles between the predicted and the original postures. The final objective function was referred to as minimum potential energy which was defined as the change in the gravity potential energy between the predicted reaching posture and the

original posture. The model was validated using the captured postures in HUMOSIM database (Human Motion Simulation Laboratory, University of Michigan), and the results showed the first objective function (minimum joint displacement) generated the most accurate reaching postures.

Hsiang and Megorrry (1997) predicted symmetric lifting motions using three optimization criteria: maximizing the smoothness of the motion pattern of the lifted load, minimizing the sudden change of the gravity centre of the human-load system, and minimizing the joint strength utilization (time integral of the square sum of the ratios between all joint moments and the corresponding joint strengths). The human body was modelled with five joint angles from elbow to ankle, and each joint angle was represented by an eight-order polynomial in time. Based on the three optimization criteria, three lifting motions and three corresponding lifted load trajectories were predicted. Participants performed the lifting tasks following the three predicted load motion patterns based on the three criteria. The peak compressive forces at the L5/S1 joint were estimated, and the results showed that the lifting motion using the first criterion produced significantly lower forces compared with the other two criteria.

Lin et al. (1999) developed a dynamic planar motion simulation (DPMS) model to predict symmetric lifting motions in the sagittal plane. The whole model consisted of three components: the trajectory model, the biomechanical model, and the optimization model. The angular displacement trajectories of five joints from ankle to elbow were formulated as polynomials of time in the trajectory model. The biomechanical model constructed the lifting dynamics so that the constraints and the objective function were represented with the coefficients of the polynomials derived from the trajectory model. Finally, the optimal joint angular trajectories were calculated in the optimization model of which the objective function was minimizing the sum of the square of the joint strength utilization (the ratio between the joint moment and the corresponding joint static strength) integrated over the movement

duration. Compared with the measured lifting motions, the simulation error was rather small, with the mean squared errors from 0.0046 rad^2 to 0.095 rad^2 .

Gündoğdu et al. (2005) simulated symmetric lifting motions in different task conditions. The human body was modelled as five-segment linkage structure including the shank, thigh, trunk, upper arm and forearm. The objective function was minimizing the time integral of sum of the square of the ratio between the predicted joint moments and the corresponding joint dynamic strengths. The joint angles were represented as seventh order polynomials in time and predicted using the genetic algorithm. Lifting motions were simulated in different task scenarios, including various lifting time durations, load weights and lifting strategies. However, no comparison between predicted and experimental motions was presented.

Abedi et al. (2012) simulated symmetric lifting motions in the sagittal plane using the optimization-based approach. The objective function was minimizing the time integral of the square of the ankle joint torque. This indicated that the CNS was assumed to maximize dynamic body balance during lifting. Results showed that the predicted lifting motion was performed in a balanced manner. However, the error between the predicted and actual motions was large, which indicated that the assumed objective function may not be an appropriate performance criterion in lifting motion prediction.

Xiang et al. (2012a) predicted three-dimensional lifting motions with different performance criteria. The whole human body was represented with a 3D skeletal linkage model with 55 degrees of freedom. Four objective functions were examined in this study. The first one was minimizing the dynamic effort defined as the time integration of the square of all joint torques. The second objective function was maximizing the balance, which was transformed to minimize the time integral of the square of the distance between zero moment point (ZMP) and foot support boundaries. The ZMP was defined as the point on the ground at which the

resultant tangential moment was zero. The third objective function was to minimize the peak low back spinal shear force, and the forth one was to minimize the peak spinal compression force at low back. The prediction results showed the maximum balance and minimum shear force performance criteria generated stoop lifting motions, while the other two performance criteria generated squat lifting motions.

In most of the motion simulation models mentioned above, a set of functions, such as polynomials of time, are pre-defined to represent the predicted motion (either hand trajectories or joint angular displacements), and the coefficients of these functions are treated as the design variables in the optimization procedures. If the task condition changes, the corresponding changes in the pre-defined functions and design variables may require significant modifications on the model. To address this problem, Chang et al. (2001) proposed a novel simulation algorithm to predict symmetric lifting motions in which no pre-determined joint angular trajectory functions were needed. In this algorithm, lifting tasks was first broken into several intervals in time, and the joint angles at the discrete points defined by these intervals were used as the design variables. The joint angular velocities and accelerations were computed with the finite difference method, and the objective function and constraints were represented as functions of these discrete angular variables. The discrete joint angles were predicted by minimizing the time integral of the sum of all the joint moments. Compared with the real lifting motions, the predicted motions were slower in the early phase of the lift, and the shoulder and elbow motions were different from the real motions in the last stage of the lift. Using this method, constraints could be added anywhere within the lifting cycle without substantially changing the model structure, which made the model more flexible and efficient.

Optimization-assisted inverse kinematics, which combines the inverse kinematics technique and the optimization procedures, is a sub-category of the optimization-based method in

human motion simulation. This type of methods facilitates the incorporation of the spatial constraints on the end-effectors in the prediction. Inverse kinematics refers to using the kinematic equations to determine the joint angles and angular velocities of a kinematic chain given the expected positions and velocities of the end-effectors (McCarthy 1990). This technique has been successfully applied in robotics (Craig 1989), and has recently attracted attentions from researchers in the area of human motion simulation.

Optimization-assisted inverse kinematics was often used to predict the upper limb reaching motions. Jung et al. (1995) developed an inverse kinematics model to predict upper limb reaching motions. In this model, the upper human body was simplified as a multi-link system consisting of four links (trunk, upper arm, lower arm, and hand) and eight degrees of freedom. To solve the kinematic redundancy, the local optimization based on the resolved motion method was used to predict the reaching motion posture by posture. The resolved motion method is an inverse kinematic technique which transforms the end-effector velocity to the joint angular velocities using a pseudoinverse of the Jacobian matrix (Whitney 1969). The objective function of the local optimization was defined as the joint range availability (JRA) which quantified the difference between the predicted and the neutral postures in the least-square form. The simulated motions were compared with the measured ones, and no significant differences in the joint angles were found at the level of significance $\alpha = 0.1$.

Using the similar method in Jung et al.'s model (1995), Jung and Shin (2010) proposed a two-handed reaching motion prediction model. The upper human body including the trunk and both arms was modelled as a seven-link system with 13 degrees of freedom. The prediction was based on two assumptions: the straight hand trajectory connecting its starting and target positions, and the bell-shaped hand velocity curve. With the resolved motion method and the local optimization, the reaching motions were simulated by a mapping which transformed the instantaneous velocity of the hand to the instantaneous joint angular

velocities. The objective function of the optimization was the same as that in Jung et al.'s model (1995): minimizing the deviation of joints from their neutral positions. Two types of reach tasks were simulated: 1) simultaneous reaching of both hands; 2) moving one hand to a target while maintaining the other one at a certain position. The simulated hand motions were compared with the experimental motions, and no statistically significant differences were found between them.

Wang (1999) proposed a geometric inverse kinematics algorithm to predict arm reach postures. The arm was model as the linkage model with seven degrees of freedom: three for shoulder, two for elbow, and two for wrist. Given the starting posture and the hand target position, the reaching postures were predicted by minimizing the sum of the norm of the joint angular velocities. The model was verified by comparing the elbow position errors between the simulated and the observed motions, and the error was less than 50 mm, which can be considered acceptable for the ergonomic application.

Dysart and Woldstad (1996) used inverse kinematics assisted by optimization to simulate the lifting final postures during symmetric lifting tasks. The human body was simplified as a five-linkage model including forearm, upper arm, torso, thigh and calf. Three performance criteria were used to predict the lifting postures separately, including minimization of the total joint toques, minimization of the joint percent strengths (the amount of torque exerted at each joint relative to the strength of the joint), and maximization of the body stability. Compared with the actual lifting trials, the prediction errors of all three criteria were significantly greater than zero, but the first objective function (minimum total torque) produced the most accurate results among the three criteria.

Zhang and Chaffin (2000) developed an optimization-based differential inverse kinematics (ODIK) approach to simulate in-vehicle reaching motions. The model used a four-segment

linkage structure with seven degrees of freedom to represent the torso, clavicle and right arm. A weighted pseudo inverse Jacobian matrix was used to relate the velocity of the end-effector (hand) and the joint angular velocities. Those weights characterized the instantaneous effort of each joint angle upon the movement, and a relative small weight indicated its corresponding joint angle had a larger contribution on the movement. The weights were assumed to be time-invariant and estimated by minimizing the time-averaged root mean square error between predicted and measured joint angular trajectories. The validation results showed this model was able to generate accurate reaching motions: the overall mean time-averaged error in joint angle was 5.2° , and the median was 4.7° .

Most of the motion simulation models reviewed above used the kinematic related variables (e.g. joint angles) as the design variables in the optimization. Actually, it is the muscle contraction that the CNS activates and controls to exert torques on human skeleton system and generate human motions. Therefore, using the muscle activation parameters as the design variables in the optimization would be a better method to uncover how the CNS plans and controls human postures and motions. Menegaldo et al. (2003) developed a biomechanical model involving muscle activities to simulate the posture control in raising-up movements from a squatting position. The model was comprised of three components: multi-body system dynamics, muscle mechanics and the geometry of the muscle moment-arms. The multi-body system dynamics constructed the relation between the joint angular movements of the human body and the muscle forces and moments exerted on the human body. The latter two components defined the relation between the muscle activities and the muscle forces and moments. The human body was simplified as a three-link inverted pendulum (shank, thigh and trunk) in the sagittal plane. The muscle activities were quantified as the excitation level from 0 to 1 in the muscle mechanics model. The whole problem was formulated as an optimization problem in which the muscle activation levels were the design variables, and the

sum of squared derivatives of the muscle forces was minimized. The simulation results showed this model was able to predict the muscular excitation patterns in postural control during rising-up tasks. However, because of the high non-linearity, numerical convergence was difficult, and the computation cost was considerably large.

2.3.1.2 Motion Simulation Using Multi-objective Optimization

The CNS may control human motions based on multiple performance criteria. For instance, Lee and Nussbaum (2013) found that both reducing low back load and enhancing dynamic body balance were important concerns in lifting motions. Therefore, using only one performance criterion (i.e. single-objective optimization) may not be sufficient to predict accurate human motions. In fact, Chang et al. (2001) indicated that even for a simple symmetric lifting task, more than one performance criterion might be necessary in human motion prediction. To overcome this limitation, multi-objective optimization (MOO) has been used in human motion simulation.

Yang et al. (2004) developed a MOO model to predict upper body postures in reaching tasks. A linkage model with 21 degrees of freedom was used to represent the torso and right arm. Three objective functions were included in the multi-objective optimization approach. The first one was joint displacement which represented the difference between the predicted and the neutral postures. The second objective function was called potential energy which represented the changes in the gravity potential energy between the predicted and the initial postures. The final one was referred to as discomfort which represented the preference of the moving sequence of different body segments. Three MOO methods, including min-max, global criterion and weighted sum methods were adopted in this study (Marler and Arora 2004). Specifically, the min-max method was to minimize the maximum value among all objective functions; the global criterion method used a weighted exponential sum of all

objective functions as the ultimate objective function, so that the MOO was transformed to a single-objective optimization; the weighted sum method was a particular form of global criterion method where the exponent was 1. All three MOO methods predicted realistic and reasonable postures, and there was no significant difference between the results from the three methods.

Xiang et al. (2010) also simulated lifting motions using the MOO method. The whole human body was represented with a linkage model with 55 degrees of freedom. Two objective functions – dynamic effort and stability – were involved to formulate the MOO problem. The dynamic effort was defined to minimize the time integral of the sum of square of all joint torques. The stability performance measure was defined to minimize the time integration of the square of the distance between zero moment point (ZMP) and foot support boundaries. The ZMP was defined as the point on the ground where the total tangential moments are zero. The weighted sum method was used in the MOO approach, and the weighting coefficients were selected from 0 to 1 with the equal interval of 0.1. Xiang et al. (2010) reported that the MOO approach did not lead to significant improvements on the simulation accuracy compared to the single-objective optimization approach which used the minimum dynamic effort as a single performance criterion.

Xiang et al. (2012b) used a hybrid optimization-based method to predict lifting motion. Different from other optimization-based models, the experimental motion data was included in the formulation of the optimization. In this study, the differences between the predicted motions and the experimental motions were referred to as tracking error, and were used as one of the two objective functions in the MOO formulated using the weighted sum method. The other objective function was dynamic effort already defined in Xiang et al. (2010). The weighting coefficients for dynamic effort and the tracking error was 0.3 and 0.7, respectively.

Compared with the previous study (Xiang et al. 2010), this hybrid MOO model generated more accurate prediction results by including the realistic experimental data

The weighting coefficients of the weighted sum method in the motion simulation models mentioned above (Yang et al. 2004; Xiang et al. 2010; Xiang et al. 2012b) were determined either arbitrarily or by using a numeration approach. In this case, finding the optimal weights was conducted using an inefficient trial-and-error principle. To address this limitation, Bottasso et al. (2006) developed a bi-level optimization method using inverse optimization to infer the weights in the weighted-sum MOO problem. In the bi-level optimization, the inner optimization was the original weighted-sum MOO problem; for the outer optimization, the design variables were the weights of the inner MOO, and the objective function was the error between the predicted results in the inner optimization and their corresponding experimental data.

Zou et al. (2012a, b) adopted the bi-level optimization approach (Bottasso et al. 2006) to determine the optimal weights of the MOO procedure in both of the seated and standing reach motion simulation. The objective function of the inner optimization was a weighted sum of the square errors in all the joint angles between the predicted and the most neutral postures. Each joint angle was assigned with one weight. These weights were determined by the outer optimization using a cost function to minimize the differences between the optimal posture predicted from the inner optimization and the real posture captured in the experiment. Participants were instructed to conduct multiple reaching tasks with different target locations, and their motions were captured. The bi-level optimization approach was adopted on every motion trial of all participants to obtain the weights of each individual trial. The final weights were calculated by averaging all the weights of every motion trail. Using these final weights, the reach postures were predicted using the one-level MOO. The results showed that the predicted postures matched the experimental data well.

2.3.2 Data-based Motion Simulation

The data-based method is another approach to solve the redundancy problem in human motion simulation. Unlike the optimization-based method mentioned above, the data-based method does not rely on pre-defined performance criteria of motor control, but the real motion data collected in experiments. Using this approach, a motion database must be constructed first, and a prediction model is developed to simulate human motions based on the collected motions. For instance, an empirical function may be constructed based on the collected motions. The function's predictors are the simulated task scenarios and human attributes, and the responses of the function are the predicted motions. In an alternative way, the collected motions can be modified to generate new motions which fit the simulated task scenarios and human attributes. Since this method predicts human motions based on real motions, just as Chaffin (2005) pointed out, the validity of the predicted motions is ensured. In this section, some data-based models are reviewed in detail.

Faraway (1997, 2000) developed a functional regression model to predict the joint angles in arm reaching motions. A quadratic regression function with functional coefficients was constructed by fitting the collected joint angular trajectories in a least-square-error manner. The predictors of the regression function were the reaching target positions, and the responses were the predicted joint angles over time. It was found that this model accounted for approximately 80% of the joint angle variations in the measured reaching motion data. With this model, both the mean and variation of the joint angular trajectories were estimated. However, since the prediction was purely in the joint angle domain, the end-effector's position at the final posture may not coincide with the target location. To solve this problem, Faraway et al. (1999) developed a rectification model using the pseudo-inverse Jacobian method (Boullion and Odell 1971). The joint angular trajectories predicted from the functional regression model were modified, so that the end-effector would land at the target

in the final posture after modification. In this model, the deviation between the original and rectified joint angles was minimized. Although the new postures satisfied the constraints, in some cases, the errors of some joint angles between the predicted and actual motions were enlarged after the rectification.

Artificial neural network (ANN) is a mathematical method which can be used for prediction. In human motion prediction, an ANN serves as a function which connects the input variables (task scenarios and human attributes) and the output variables (joint motions and postures). The coefficients of the ANN are calculated and updated based on the existing motion data in the learning phase. Jung and Park (1994) developed a neural network model based on a back propagation learning method to predict arm reaching postures. The input of this ANN model consisted of the target coordinates and anthropometric variables (e.g. gender), and the output was the three-dimensional motion trajectories of the right arm joint centres (shoulder, elbow and wrist). The measured motions were divided into two parts: the training motions and the testing motions. The training motions were used to update the weighting coefficients in the neural network. The simulation results were compared with the testing motions, and no significant difference was between the predicted and the real motion trajectories.

Perez and Nussbaum (2008) developed a neural network model to predict manual lifting motions. The task conditions (e.g. target locations, movement durations) and human attributes (e.g. joint strengths) were input to the model, and the outputs of the model were the joint angular trajectories of lifting motions. Both the static postures and dynamic motions of lifting tasks in the symmetric sagittal plane and three-dimensional space were predicted. The model was trained using subsets of an existing lifting motion database, and the remaining motion data was used to evaluate the prediction performance of the model. The prediction error was in the range of 5°-20° per joint, which was similar in magnitude of the within-subject variation in the motion database.

In the data-based models mentioned above, the real motion data was used to construct an empirical function (e.g. regression function, ANN) to predict motions in novel scenarios. Alternatively, human motions can be simulated by modifying and editing the real motions so that the predicted motions can satisfy new task scenarios. This motion editing technique is commonly used in computer graphics to generate novel human motions. In general, a pre-captured motion is modified to fit the required task scenarios while the basic structure of the original motion is preserved.

Witkin and Popovic (1995) proposed a motion warping technique which can generate a family of new motions based on a pre-existing motion. Each frame of the simulated motion was transformed from its corresponding frame of the original motion by certain mapping functions to ensure the simulated motion can satisfied the new task constraints on a set of key frames. This motion warping technique had two main limitations. First, it was a purely geometric technique without considering the physical properties of human motions. Therefore, some motions generated by this technique were prone to be distorted and unnatural. Secondly, the constraints on the key frames were applicable in this model, but enforcing constraints between the key frames or over a period of time were difficult.

Gleicher (1997) developed a motion editing method based on spacetime constraints (Witkin and Kass 1988). The joint angle trajectories were represented as parametric curves, such as B-Spline curves. The entire problem was transformed into a typical constrained optimization that minimized the difference between the original motion and the motion after editing. In this model, unlike the motion warping technique (Witkin and Popovic 1995), the constraints over a range of time were applied. The performance of the model showed it was able to solve the numerical constrained optimization sufficiently fast so that real time interactive motion editing was permitted.

Park et al. (2004, 2005a, 2005b, 2008) proposed a memory-based motion simulation (MBMS) model to predict human motions by modifying the real stored motions. This model was comprised of three components: a motion database, a motion search method, and a motion modification algorithm. The motion database stored joint angular displacements of real motions and served as the memory of past motions. According to generalized motor program (GMP) theory (Schmidt and Lee 1999), the past motions in human beings' memory were used as templates for motion planning when humans performed novel but similar tasks. The simulation started with a submission of a simulated scenario to the model, and the motion search method searched the motion database to retrieve motions which had similar scenarios as the simulated one (none or many could be found). These retrieved motions were called "root motions". Each root motion was further analysed and each joint angular trajectory of the root motion was assigned with a symbolic string according to its monotonic interval sequence. Three symbols were used: 'U' for monotonically increasing, 'D' for monotonically decreasing, and 'S' for stationary. The root motion was modified in the motion modification algorithm in which the boundary points of these monotonic intervals were adjusted to fit the new task constraints while the sequence of monotonic intervals (the sequence of the symbols in the string) remained the same. The model was evaluated in various types of manual tasks (e.g. reaching and load-transfer task), and the errors between the simulated and real motions were comparable to the motion variability under identical task conditions.

Based on the MBMS model, Park et al. (2006) further developed a reaching posture prediction model in the presence of obstructions. In this model, the whole workplace was partitioned into numerous small regions called cells. Then, the recorded postures reaching the target cell were retrieved from the database. Finally, the model selected the postures which avoided the obstructions from the retrieved ones and modified them so that they matched the

constraints and targets exactly. The simulation results showed the model was able to plan the reaching postures rapidly and robustly.

Rosenbaum et al. (1995) developed a reaching motion prediction model based on the stored reaching postures. The prediction process was separated into two steps: planning and execution. In the planning step, each stored reaching posture was assigned with a weight according to two costs: the spatial error cost which reflected the distance between the simulated target and the target of the stored posture, and the travel cost determined by how costly it was to move from the starting posture to the stored posture. All the stored postures were treated as vectors whose components were the joint angles. Therefore, the simulated reaching posture was computed as a weighted sum of these vectors. In the execution step, each joint angular motion was predicted by connecting initial and final postures with an S-shaped function which generated bell-shaped angular velocity profiles. The performance of this model was good, since there was no significant difference between the simulated and actual postures.

Rosenbaum et al. (2001) further developed the previous posture based model (Rosenbaum et al. 1995) to predict the reaching and grasping motions in which obstacles were avoided. A constraint hierarchy – a set of task requirements ordered in their priority, was first established. The prediction of reaching postures was a two-stage process: selecting the most promising posture from the stored candidate postures, and then modifying the selected posture to fit the constraints. Here, the most promising posture referred to the one best matching the constraint hierarchy in the collected posture database. After the reaching posture was acquired, the similar execution process as that in the previous model (Rosenbaum et al. 1995) was carried out to generate the motion connecting the starting and reaching postures. The prediction results were more accurate at the starting and ending movement periods compared with the

middle part of the motion, since the mean squared errors (MSE) between the simulated and real data were smaller at the starting and ending parts than the middle part.

2.3.3 Limitations on the Existing Human Motion Simulation Models

The major advantage of the optimization-based methods is that no database of realistic motions is needed. Also, these methods may provide insights into the mechanisms by which the CNS plans human motions. However, there are still some limitations with the optimization-based methods.

The essential limitation of the optimization-based methods is that there is no evidence for validating the assumption regarding the performance criteria in the optimization procedure. If the assumption regarding the performance criteria is invalid, the predicted motions will be invalid as well. Although the inverse-optimization approach (Bottasso et al. 2006) is able to infer the weights in the weighted-sum MOO for motion simulation, the objective functions in the MOO are still based on pre-assumptions.

In addition, the performance criteria might be task-related or individual specific. In other words, there is no universal performance criterion which is applicable for all task scenarios and different individuals. Another limitation of the optimization-based method is that only one human motion trajectory can be predicted for a certain task scenario. Actually, the human motion is a stochastic process. People may choose different motion patterns for the same task. This variation phenomenon in human motions is neglected in the optimization-based methods. Finally, the optimization procedures in most of the models are computationally intensive, which makes it difficult to apply these methods for real-time analysis and simulation.

The data-based methods do not rely on certain performance criteria but on the real motion data, so the main advantage of the data-based methods is the intrinsic realism of the simulated motions which are derived from the actual motions. However, the data-based methods have

their own inherent limitations. First, these methods require collecting actual motions for each simulated task, which are time-consuming and expensive. Second, the motion prediction heavily relies on the recorded motions. If the simulated motion scenario is far different from the scenarios of the stored motions, the prediction accuracy declines significantly. For instance, Faraway (2003) pointed out that, in the reaching motion prediction, when the targets in simulation scenario were located far from the targets of the recorded motions, the prediction accuracy would be deteriorated significantly. Third, most of the empirical models are pure kinematic models without considering the kinetic properties of the human motion, so biomechanical analysis of human motions based the data from these models will be insufficient.

Besides these, there are several common limitations shared by both the data- and optimization-based methods. First, many factors accounting for inter-individual differences (e.g. age and gender) are neglected in the existing models. Besides, the motion durations are pre-defined, while in reality it is difficult to accurately estimate the motion duration before the motion execution.

To overcome some of these limitations (e.g., low validity of the predicted motions from the optimization-based methods, unable to account for the effects of inter-individual differences on simulated motions), two novel hybrid models which combine the advantages of the optimization-based and data-based methods were developed for lifting motion simulation. Specifically, single-objective optimization approach was used for the model presented in Chapter 7; and in Chapter 8, a hybrid model using multi-objective optimization was further developed based on the single-objective-optimization model.

Chapter 3 Experimental Protocol and Data Collection

3.1 Participants

Eleven younger participants (six males and five females) aged between 20 – 30 years old and twelve older participants (seven males and five females) aged over 55 years old were recruited from the university and local community. All were free from any musculoskeletal disorders in the last six months. Their demographic information including their age, height, body mass and maximum lifting capacity (MLC) is listed in Table 3.1. Before the experiment, each participant signed the informed consent form approved by the Nanyang Technological University IRB.

Table 3.1 Demographic information of participants

	Young		Old	
	Mean	SD	Mean	SD
Age (years)	23.6	2.0	66.1	5.6
Height (cm)	167.0	6.9	163.4	9.2
Body Mass (kg)	54.6	7.2	61.2	10.4
MLC (kg)	26.0	9.3	21.8	11.6
MLC SD between repetitions (kg)	2.7	1.5	3.0	1.6

3.2 Apparatus

An eight-camera optoelectronic motion capture system (Motion Analysis Eagle System, Santa Rosa, CA, USA) was used to measure the whole-body kinematics during lifting. Two force plates (AMTI OR6-7, Watertown, MA, USA) were used to measure ground reaction forces. The motion capture system and the force plates were synchronized and set at the sampling rates of 100 Hz and 1000 Hz, respectively. A commercial dynamometer (Biodex System 4 Pro, Shirley, NY, USA) was used to measure the MLC of each participant.

3.3 Experimental Procedures

At the beginning, participants carried out a warm-up session involving a 2-min jogging on a treadmill and whole-body stretching exercises. Participants then conducted an isokinetic lifting test to measure their MLC. Specifically, participants performed 6 groups of isokinetic lifting tasks, and each group contained 5 continuous repetitions. In each repetition, participants lifted a bar of a lift simulation attachment connected with the dynamometer from the knee to elbow height (Fig. 3.1). They were instructed to use a squat strategy during lifting, and a 2-min rest was given after each lifting group. The velocity of the bar was set at 78 cm/s, which was consistent with Timm (1988). The first two lifting groups were practice sessions in which participants were instructed to use about 50% of their maximum strength. In the last four lifting groups, participants were instructed to use their maximum strength, and the average peak force of these four lifting groups was defined as the MLC for each participant.

Figure 3.1 Isokinetic test for the measurement of MLC

After the MLC exercise, 31 reflective markers were placed on the selected body landmarks of each participant (Table 3.2). During the lift task, participants lifted loads from the floor to one of two shelves (Fig. 3.2). One shelf (symmetric) was placed in the sagittal plane, and the other one (asymmetric) was placed at 60° to the right of the sagittal plane. The height of the shelves was adjustable to provide different destination heights. The height of the symmetric shelf was set at the wrist, elbow and shoulder heights of each participant during erect stance, and the height of the asymmetric shelf was set at the wrist height only. The distance from the edge of the shelves to the participants' standing point (i.e. the middle point of the two ankle joints) was 58 cm. The lifted load was a square box (length $\times$ width $\times$ depth: 0.34m $\times$ 0.24m $\times$ 0.26m) with two handles on its sides. The box was placed right before participants prior to each lift, and the initial horizontal distance from the box centre to participants' ankles was 40 cm. Three levels of load weights (5%, 15% and 25% MLC of each participant) were involved in the lifting task. Therefore, there were totally twelve task conditions (four load destinations $\times$ three load weights) for each participant. Participants performed three repetitions in each task condition which were randomly ordered. Before each lifting trial, participants were informed of the lifted load weight and the destination. During lifting, participants stood on the force plates (with each foot on one force plate), and lifted the box by holding its handles. They were instructed to use freestyle lifting strategies and self-selected lifting speeds. Participants were not allowed to move their feet during lifting, although they could lift the heel of the left foot when transferring the box to the asymmetric shelf. Prior to the data collection, participants were provided with a practice session to get familiar with the lifting protocol. To minimize any confounding effects caused by fatigue, a 30-second break (standing without load) was given after every lifting trial. The specific dependent variables, data analysis techniques and statistical analysis procedures are outlined in the subsequent chapters.

Table 3.2 Body landmarks for reflective marker placement

Marker No.	Body landmarks
1	Forehead
2	Chin
3	Suprasternal
4	C7/T1
5 and 6	Right and left acromions
7 and 8	Right lateral and medial epicondyles of humerus
9 and 10	Left medial and lateral epicondyles of humerus
11 and 12	Right styloid processes of radius and ulna
13 and 14	Left styloid processes of ulna and radius
15 and 16	Right and left ASIS (anterior superior iliac spines)
17 and 18	Right and left PSIS (posterior superior iliac spines)
19	L5/S1
20 and 21	Right lateral and medial epicondyles of femur
22 and 23	Left medial and lateral epicondyles of femur
24 and 25	Right lateral and medial malleoli
26 and 27	Left medial and lateral malleoli
28 and 29	Right and left heels
30 and 31	Right and left 5th metatarsal

Figure 3.2 A participant performing symmetric (left) and asymmetric (right) lifting trials

Chapter 4 Effects of Age and its Interaction with Work-related Factors on Biomechanics during Symmetric Lifting

4.1 Introduction

Low back pain (LBP) is a highly prevalent and costly musculoskeletal disorder (Katz 2006; Lawrence et al. 1998). LBP incidence is often associated with manual lifting (Hoogendoorn et al. 2000; Kuiper et al. 2005). Thus, in order to well address LBP problems, lifting biomechanics has been studied under various conditions in the existing literature (reviewed in Section 2.2).

LBP is more prevalent among older adults (Holmstrom and Engholm 2003; Peek-Asa et al. 2004; Heiden et al. 2013). Given that the workforce is aging worldwide, there is a need to study age-related differences in lifting biomechanics. As discussed in Section 2.2.3, the only studies that investigated age effects on lifting biomechanics were conducted by Shin et al. (2006) and Burgess et al. (2009). There were three major limitations with Shin et al. (2006) and Burgess et al. (2009):

1. They did not report low back load which is a crucial contributor to LBP incidence (Norman et al. 1998);
2. Age-related differences were only investigated in hip and trunk kinematics instead of whole-body kinematics;
3. The interaction effects between age and some important work-related factors (i.e., load weight, asymmetric destination) have not been studied.

In order to address these limitations, two major purposes were proposed in the present study: 1) to examine the age effects on lifting biomechanics, including low back load, whole body starting and ending postures, and trunk, hip and knee kinematics; and 2) to determine the interaction effects of age and work-related factors (i.e. load weight and destination height) on lifting biomechanics. It was hypothesized that: 1) age would significantly affect lifting kinematics and low back load; and 2) there would be significant interaction effects between age and the selected work-related factors (i.e. load weight and destination height) on lifting biomechanics.

4.2 Methods

The symmetric lifting trials collected in the lifting experiment (Chapter 3) were analysed in this study to examine the effects of age, load weight and destination height on lifting biomechanics.

4.2.1 Dependent Measures

The dependent measures included the joint angles of the starting and ending body postures, the peak angular velocities and accelerations of the trunk, hip and knee, and the peak and average low back moments. The selected kinematic measures well account for the lifting posture and motion patterns, and have been extensively studied in previous researches (Allread et al. 1996; Chen 2000; Sparto et al. 1997). Low back moment is a primary indicator of the LBP risk in lifting tasks (Faber et al. 2011; Hoozemans et al. 2008; Kingma et al. 2010). In addition, the postural index (PI) defined as the ratio between the knee flexion and the sum of ankle, hip and trunk flexion in the starting posture, are used to quantify the lifting strategy (Burgess and Abernethy 1997).

The definition of the joint centres was given by Table 4.1 (Rab et al. 2002). Since only symmetric lifting tasks were analysed in this study, the joint angles were defined in the sagittal plane using a 2-D rigid linkage model (Fig. 4.1). In this linkage model, the low back joint centre was at the L5/S1 location, and the other joint centers were defined as the middle points of the corresponding bilateral joint centres. The reference angle (0°) for each joint was defined at the standard anatomical position. Positive angles indicate shoulder extension, ankle dorsiflexion, and flexion of the elbow, trunk, hip and knee.

Table 4.1 Definitions of joint centres

Joint centers	Definitions
C7/T1	C7/T1 (4)
Right shoulder	17% of the distance between right (5) and left (6) acromions inferiorly relative the right acromions (5)
Left shoulder	17% of the distance between right (5) and left (6) acromions inferiorly relative the left acromions (6)
Right elbow	Midpoint of the right lateral (7) and medial (8) epicondyles of humerus
Left elbow	Midpoint of the left lateral (10) and medial (9) epicondyles of humerus
Right wrist	Midpoint of the right styloid processes of radius (11) and ulna (12)
Left wrist	Midpoint of the left styloid processes of radius (14) and ulna (13)
Low back	L5/S1 (19)
Right hip	18% of the distance between two ASISs (15 and 16) posteriorly, 34% of the same distance inferiorly relative to right ASIS (15)
Left hip	18% of the distance between two ASISs (15 and 16) posteriorly, 34% of the same distance inferiorly relative to left ASIS (16)
Right knee	Midpoint of the right lateral (20) and medial (21) epicondyles of femur
Left knee	Midpoint of the right lateral (23) and medial (22) epicondyles of femur
Right ankle	Midpoint of the right lateral (24) and medial (25) malleoli
Left ankle	Midpoint of the left lateral (27) and medial (26) malleoli

* The numbers in the parentheses are the serial numbers for the reflective markers in table 3.2.

Figure 4.1 The rigid linkage model in the sagittal plane

Low back moment was calculated by a 3-D inverse dynamics model proposed by Kingma et al. (1996). The inputs of the inverse dynamics model were the ground reaction forces, and the body segment kinematics and anthropometry. The parameters for estimating the segment anthropometry (i.e., segment mass, center of mass and moment of inertia) were obtained according to de Leva (1996). The reactive forces and moments at the intersegment joints were calculated by analyzing each body segment in a bottom-up sequence (i.e., foot-shank-thigh-pelvis). The dynamic equations of motion for each body segment were described by Kingma et al. (1996):

$$\sum_{k=1}^p \mathbf{F}_k + m\mathbf{g} = m\mathbf{a} \quad (\text{Eq. 4.1})$$

$$\sum_{k=1}^p ((\mathbf{v}_k - \mathbf{v}_{COM}) \times \mathbf{F}_k) + \sum_{l=1}^q \mathbf{M}_l = d(\mathbf{I}\boldsymbol{\omega})/dt \quad (\text{Eq. 4.2})$$

where $\mathbf{F}_k$ are the k -th vector of the external and joint reactive forces on the body segment; p is the number of external and joint reactive forces; $\mathbf{g}$ is the gravity vector; $\mathbf{v}_k$ is the vector for the application point of $\mathbf{F}_k$; $\mathbf{v}_{COM}$ is the vector for the segment COM; $\times$ is the vector product; $\mathbf{M}_l$ is the l -th joint moment on the segment; q is the number of joint moments; m , $\mathbf{a}$, $\mathbf{I}$ and $\boldsymbol{\omega}$ are the mass, the linear acceleration vector, the inertia tensor and the angular velocity vector of the segment, respectively; $d(\dots)/dt$ is the time derivative of the expression within the parentheses. Since only symmetric lifting was performed in this study, the flexion-extension component of the calculated 3-D low back moment vector was used to assess low back moment.

A lifting trial can be separated into two phases: the lifting phase and deposit phase (Gagnon and Smyth 1992; Hsiang et al. 1998). The lifting phase was defined as the period between the start of lifting and the time when the box was closest to the trunk (closest-to-body-time), and mainly involved the extensions of the trunk and lower extremities. The deposit phase started

from the closest-to-body-time and lasted till the end of the lift, and mainly involved elbow and shoulder extensions and trunk flexion. The peak low back moment was defined on the whole lifting period, and the average low back moment was calculated for these two phases, separately.

4.2.2 Statistical Analysis

The independent variables were age, load weight, and destination height. Since each participant conducted three repetitions of lifting within each task condition, the statistical analysis was performed by averaging the dependent measures of the three repetitions. Because of the large number of dependent variables, a multivariate ANOVA (MANOVA) was conducted to control the experiment-wise error rate. Then, a mixed ANOVA with repeated measures was conducted to test the effects of the independent variables and their interactions which showed significant effects in MANOVA. Post-hoc analysis was conducted on the dependent variables that were significantly affected by destination height and load weight using the Bonferroni test. In case of a significant interaction between age and any one of the selected work-related factors (i.e. destination height or load weight), between-subject t-test was applied to test the effects of age for each level of the work-related factor, and Bonferroni test was used to examine the effects of the work-related factor within each age group, separately. Level of significance was set at 0.05. Before analysis, the assumptions of normality of residuals, homogeneous variance of residuals, and independence of observations were tested using the approaches suggested by Montgomery (2005). The test revealed a violation on the assumption of homogeneous variance of residuals. A natural logarithmic transformation was performed on the dependent variables which violated the assumption, and the violation was remedied after the transformation. The sphericity assumption of the repeated measures ANOVA was evaluated using Mauchly's sphericity test, and Greenhouse-Geisser adjustment was adopted in case of a violation.

4.3 Results

The results of the MANOVA showed that the main effects of the three factors (i.e., age, load weight, and destination height) and the two-way interactions between age and the two work-related factors were significant (i.e., p -values < 0.05). The main and interaction effects on dependent variables revealed by ANOVAs were presented below.

4.3.1 Age Effects

The average PI in older participants was 0.54 (SD = 0.91), ranging from 0.33 to 0.71. The average PI in younger participants was 0.60 (SD = 0.91), ranging from 0.38 to 0.76. This indicated that both age groups adopted a lifting strategy intermediate between full stoop (PI = 0.11) and full squat (PI = 0.80), which can be considered as a semi-squat strategy (Burgess 2003).

Significant age effects were found in the starting and ending postures as well as the trunk and knee angular velocities and accelerations (Table 4.2). In particular, older participants showed significantly lower trunk and knee flexion, but higher hip flexion at the starting posture compared with younger participants. For the ending posture, shoulder and hip joint angles were significantly higher in older participants. Significantly lower peak trunk and knee angular velocities and accelerations were observed in older participants as well.

No significant differences between two age groups were found in the peak low back moment and the average low back moment in the lifting phase. However, older participants experienced significantly higher average low back moment in the deposit phase compared with their younger counterparts (Table 4.2).

4.3.2 Destination Height Effects

Destination height significantly affected the starting and ending postures (Table 4.3). In the starting posture, elbow and hip flexion increased with higher destination, and knee flexion and ankle dorsiflexion reached the highest values at the elbow destination height. In the ending posture, increased destination height resulted in larger upper-extremity extensions and a more erect body posture. In particular, shoulder extensions increased, and trunk, hip and knee flexion decreased with increased destination height. People tended to slow down lower extremity motions but accelerate trunk extension with higher destination height, because the peak trunk angular velocity and acceleration significantly increased, and the peak knee angular velocity and acceleration significantly decreased with increased destination height. In addition, increased destination height resulted in a significant increase in the peak low back moment and a significant decline in the average low back moment in both the lifting and deposit phases.

4.3.3 Load Weight Effects

Increased load weight resulted in a significant increase in the low back moment (Table 4.4). Hip flexion in the starting posture increased significantly with higher load weight. In the ending posture, increased load weight brought about significant increases in shoulder extension and flexion of the elbow, trunk and hip. Therefore, higher load weight resulted in more forward leaning body postures at both the start and end of the lifting task. In addition, significant decreases were found in the peak trunk angular velocity and peak hip angular acceleration with increased load weight.

Table 4.2 Main effects of age on dependent variables

	Age				p-value	F-value
	Young		Older			
	Mean	SD	Mean	SD		
<i>Starting Posture</i>						
Elbow joint angle (°)	11.13	12.37	7.59	6.10	0.749	0.105
Shoulder joint angle (°)	56.13	11.31	63.25	10.45	0.129	2.501
Trunk angle (°)	44.69	6.65	30.56	5.82	<0.001	28.573
Hip joint angle (°)	81.72	5.65	88.33	7.54	0.030	5.441
Knee joint angle (°)	89.91	17.43	73.82	18.87	0.048	4.403
Ankle joint angle (°)	21.88	6.61	15.78	8.59	0.073	3.550
<i>Ending Posture</i>						
Elbow joint angle (°)	18.81	10.33	16.81	6.08	0.493	0.488
Shoulder joint angle (°)	83.13	21.40	89.79	19.45	0.042	4.712
Trunk angle (°)	8.00	6.71	6.09	5.70	0.214	1.640
Hip joint angle (°)	10.70	7.28	16.76	8.06	0.009	8.304
Knee joint angle (°)	3.28	7.25	2.31	3.73	0.758	0.098
Ankle joint angle (°)	-3.35	3.64	-5.14	2.62	0.079	3.397
<i>Peak Angular Velocities and Accelerations</i>						
Peak trunk extension velocity (°/s)	66.51	16.84	35.72	10.13	<0.001	35.518
Peak hip extension velocity (°/s)	126.03	20.77	111.26	24.38	0.124	2.575
Peak knee extension velocity (°/s)	138.71	31.44	91.23	27.03	<0.001	17.844
Peak trunk acceleration (°/s ²)	217.43	62.63	116.22	47.10	<0.001	23.801
Peak hip acceleration (°/s ²)	390.71	99.30	316.50	116.51	0.096	3.037
Peak knee acceleration (°/s ²)	524.84	159.49	322.59	120.55	0.001	13.678
<i>Low Back Moment</i>						
Peak moment (Nm)	165.01	33.38	163.91	38.96	0.942	0.005
Average moment – Lifting Phase (Nm)	116.16	26.89	122.24	30.76	0.578	0.319
Average moment – Deposit Phase (Nm)	63.85	19.90	89.81	30.03	0.002	12.242

Table 4.3 Main effects of destination height on dependent variables

	Destination Height						p-value	F-value
	Wrist Height		Elbow Height		Shoulder Height			
	Mean	SD	Mean	SD	Mean	SD		
<i>Starting Posture</i>								
Elbow joint angle (°)	8.53 ^A	9.42	8.01 ^A	9.33	11.31 ^B	10.31	<0.001	16.381
Shoulder joint angle (°)	60.81	10.44	59.47	11.68	59.25	12.15	0.083	2.645
Trunk angle (°)	37.28	9.49	37.11	9.56	37.57	9.34	0.144	2.029
Hip joint angle (°)	84.62 ^A	7.55	85.13 ^{A, B}	7.43	85.75 ^B	7.45	0.002	7.601
Knee joint angle (°)	80.32 ^A	19.72	83.03 ^B	19.46	81.20 ^{A, B}	20.57	0.028	3.886
Ankle joint angle (°)	18.71 ^{A, B}	8.15	19.21 ^A	8.34	18.17 ^B	8.42	0.049	3.248
<i>Ending Posture</i>								
Elbow joint angle (°)	18.11 ^A	9.03	21.07 ^B	8.30	14.12 ^C	6.32	<0.001	35.731
Shoulder joint angle (°)	69.94 ^A	11.04	78.13 ^B	10.82	111.75 ^C	7.31	<0.001	450.34
Trunk angle (°)	13.13 ^A	5.25	5.91 ^B	3.65	1.97 ^C	3.60	<0.001	189.71
Hip joint angle (°)	20.69 ^A	8.39	12.35 ^B	6.02	8.55 ^C	4.65	<0.001	78.013
Knee joint angle (°)	4.42 ^A	8.06	2.60 ^A	4.08	1.30 ^B	3.43	0.013	6.487
Ankle joint angle (°)	-5.64 ^A	3.86	-4.05 ^B	2.73	-3.16 ^C	2.58	<0.001	19.720
<i>Peak Angular Velocities and Accelerations</i>								
Peak trunk extension velocity (°/s)	44.93 ^A	18.80	50.35 ^B	19.76	56.07 ^C	21.98	<0.001	58.705
Peak hip extension velocity (°/s)	116.08	23.63	119.66	24.66	119.23	23.44	0.073	2.784
Peak knee extension velocity (°/s)	123.32 ^A	38.15	116.76 ^B	36.19	101.73 ^C	35.68	<0.001	52.589
Peak trunk acceleration (°/s ²)	154.89 ^A	70.43	158.47 ^A	66.96	180.50 ^B	84.19	0.001	9.113
Peak hip acceleration (°/s ²)	351.39	123.93	358.55	116.26	346.04	103.91	0.323	1.122
Peak knee acceleration (°/s ²)	443.14 ^A	178.13	424.73 ^A	167.51	390.07 ^B	171.27	0.002	9.657
<i>Low Back Moment</i>								
Peak moment (Nm)	162.64 ^A	35.60	164.55 ^{A, B}	36.28	166.12 ^B	37.49	0.002	7.092
Average moment – Lifting Phase (Nm)	122.15 ^A	28.00	118.09 ^B	29.70	117.77 ^B	29.67	<0.001	12.331
Average moment – Deposit Phase (Nm)	92.89 ^A	29.63	76.26 ^B	26.33	63.02 ^C	21.82	<0.001	244.85

* The conditions with the same superscripted letter (A, B or C) were not significantly different from each other.

Table 4.4 Main effects of load weight on dependent variables

	Load weight						p-value	F-value
	5% MLC		15% MLC		25% MLC			
	Mean	SD	Mean	SD	Mean	SD		
<i>Starting Posture</i>								
Elbow joint angle (°)	8.66	9.02	9.34	9.20	9.85	11.03	0.316	1.183
Shoulder joint angle (°)	59.63	11.62	59.76	10.97	60.14	11.79	0.591	0.533
Trunk angle (°)	37.10	9.48	37.38	9.24	37.49	9.66	0.282	1.305
Hip joint angle (°)	84.58 ^A	7.50	85.09 ^A	7.39	85.83 ^B	7.54	<0.001	10.244
Knee joint angle (°)	80.76	20.18	81.34	19.27	82.44	20.37	0.099	2.440
Ankle joint angle (°)	18.63	8.50	18.76	8.06	18.71	8.37	0.934	0.059
<i>Ending Posture</i>								
Elbow joint angle (°)	15.47 ^A	6.75	17.72 ^B	7.30	20.11 ^C	10.25	<0.001	14.274
Shoulder joint angle (°)	84.20 ^A	20.60	86.69 ^B	20.72	88.92 ^B	20.59	0.001	10.976
Trunk angle (°)	5.93 ^A	5.80	7.25 ^B	5.87	7.83 ^B	6.98	<0.001	23.219
Hip joint angle (°)	12.34 ^A	7.42	14.00 ^B	7.78	15.25 ^B	9.31	<0.001	13.137
Knee joint angle (°)	2.63	4.18	2.83	4.43	2.87	7.83	0.196	1.771
Ankle joint angle (°)	-3.35 ^A	2.58	-4.33 ^B	2.72	-5.17 ^B	4.06	<0.001	14.817
<i>Peak Angular Velocities and Accelerations</i>								
Peak trunk extension velocity (°/s)	51.63 ^A	20.88	50.90 ^A	21.00	48.81 ^B	20.22	0.004	7.717
Peak hip extension velocity (°/s)	119.09	22.69	119.26	24.39	116.63	24.69	0.261	1.381
Peak knee extension velocity (°/s)	112.77	38.36	115.83	37.68	113.21	37.28	0.219	1.574
Peak trunk acceleration (°/s ²)	167.27	76.14	164.25	75.37	162.34	73.71	0.446	0.756
Peak hip acceleration (°/s ²)	363.59 ^A	107.68	355.30 ^A	122.49	337.09 ^B	113.06	0.034	3.669
Peak knee acceleration (°/s ²)	429.82	169.40	419.88	175.56	408.23	175.62	0.151	1.975
<i>Low Back Moment</i>								
Peak moment (Nm)	151.99 ^A	31.81	165.26 ^B	34.18	176.06 ^C	38.99	<0.001	53.192
Average moment – Lifting Phase (Nm)	103.29 ^A	20.68	120.64 ^B	25.67	134.08 ^C	31.44	<0.001	81.218
Average moment – Deposit Phase (Nm)	62.77 ^A	22.69	78.85 ^B	25.69	90.56 ^C	30.53	<0.001	99.773

* The conditions with the same superscripted letter (A, B or C) were not significantly different from each other.

4.3.4 Interaction Effects between Age and Destination Height

The results showed significant interaction effects of age and destination height on the final shoulder extension ($p = 0.042$, $F = 3.412$), the average low back moment in the lifting phase ($p = 0.01$, $F = 5.168$) and the average low back moment in the deposit phase ($p = 0.014$, $F = 4.777$) (Fig.4.2). When investigating the effects of destination height within each age group, it was found that the final shoulder extension increased significantly with increased destination height in both young and older adults (Fig. 4.2-a). However, when examining the age-related differences in the final shoulder extension at each destination height level, significant age effects were found at the wrist and elbow destination height levels only. In particular, compared with younger participants, older participants had significantly larger shoulder extension at both levels.

The results showed no significant differences in the average low back moment in the lifting phase between any two levels of destination height for younger participants (Fig. 4.2-b). However, older participants showed significant decreases in the average low back moment in the lifting phase from wrist to elbow level as well as from wrist to shoulder level. In addition, no significant age-related differences in the average low back moment in the lifting phase were found in any destination height level.

The average low back moment in the deposit phase became significantly smaller in each age group when destination height increased (Fig. 4.2-c). However, the average moment of older participants seemed to decrease with increased destination height in a different way from that of younger participants. Among older participants, the average low back moment in the deposit phase decreased similar amounts from the wrist level to elbow level (15.98 Nm) as from the elbow level to shoulder level (17.24 Nm). Among younger participants, however, decreases of the average moment from the wrist level to elbow level and from the elbow level

to shoulder level were quite different (17.34 Nm vs. 8.87 Nm). We also found that older participants experienced significantly higher average moment than did younger participants at each destination height level.

Figure 4.2 Interaction effects between age and destination height. The solid lines and dash-dot lines correspond to data from younger and older participants, respectively. ‘*’ indicates significant age effects. ‘x’ indicates significant destination height effects. Error bars indicate 1 standard error.

4.3.5 Interaction Effects between Age and Load Weight

Significant age and load weight interaction effects (Fig. 4.3) were found on the initial trunk flexion ($p = 0.02$, $F = 4.284$), the final elbow flexion ($p = 0.037$, $F = 4.288$), trunk flexion ($p = 0.011$, $F = 5.064$) and hip flexion ($p = 0.01$, $F = 5.204$), and the average low back moment

in the deposit phase ($p = 0.039$, $F = 4.003$). The results showed no significant effects of load weight on the initial trunk flexion in both age groups (Fig. 4.3-a). However, the mean value of the initial trunk flexion changed among three load weight levels in different manners between younger and older groups. In the younger group, the mean initial trunk flexion remained the same from 5% to 15% MLC level, and increased from 15% to 25% MLC level; in the older group, however, the mean initial trunk flexion first increased from 5% to 15% MLC level, then decreased from 15% to 25% MLC level. Also, older people adopted significantly smaller initial trunk flexion compared to their younger counterparts within each load weight condition.

The results showed significant increases in the final elbow flexion from 5% to 15% MLC level and from 5% to 25% MLC level in both age groups (Fig 4.3-b). Younger participants showed larger increases in the mean values of the final elbow flexion as load weights increased (an increment of 7.3° from 5% to 25% MLC level) compared with older participants (an increment of 2.16° from 5% to 25% MLC level). In addition, there were no significant age-related differences in the final elbow flexion within any load weight level.

The final trunk flexion increased significantly from 5% to 15% MLC level as well as from 15% to 25% MLC level in younger participants (Fig. 4.3-c). In older participants, however, no significant changes in the final trunk flexion were found between 15% and 25% MLC levels. In addition, the results showed no age-related significant differences in the final trunk flexion within any of the three load weight levels.

For the final hip flexion, significant differences between varied load weight levels were only discovered in younger participants (Fig. 4.3-d). In particular, the final hip flexion increased significantly from 5% to 15% MLC level as well as from 5% to 25% MLC level. In addition,

older participants showed significantly higher final hip flexion within 5% and 15% MLC levels compared with younger participants.

The results showed significant increases in the average low back moment in the deposit phase with increased load weights in both age groups (Fig 4.3-e). Older participants showed significantly higher average low back moments in the deposit phase than younger participants within every load weight level. However, based on the mean value of the moments in each load weight level, the age-related differences seemed to decrease with increased load weight. Specifically, the differences in the mean value of the moments between two age groups were 26.67, 26.42, and 24.79 Nm at 5%, 15% and 25% MLC levels, respectively.

Figure 4.3 Interaction effects between age and load magnitude. The solid lines and dash-dot lines correspond to data from younger and older participants, respectively. ‘*’ indicates significant age effects. ‘x’ indicates significant load magnitude effects. Error bars indicate 1 standard error.

4.4 Discussion

The main purpose of the present study was to investigate the effects of age and its interactions with work-related factors (i.e. destination height and load weight) on lifting biomechanics. Participants were divided into two groups according to their age and were instructed to perform lifting tasks that were defined by different combinations of destination heights and load weights. Lifting biomechanics were assessed by using the measures of starting and ending postures of lift, lower-extremity joint kinematics, and low back moment.

4.4.1 Age Effects

The starting posture of lift is able to characterize the lifting strategy (Burgess and Abernethy 1997), and has important impacts on low back load and risks of LBP (Burgess 2003; Hsiang, et al. 1997). In this study, both younger and older participants adopted a lifting strategy (i.e. semi-squat strategy) between full stoop and full squat based on the postural index (PI) values. This is consistent with previous studies (Albert et al. 2008; Burgess and Abernethy 1997; Burgess 2003) which reported people tended to use a semi-squat strategy when they were

instructed to use self-selected lifting strategies to lift low lying loads. Older participants showed significantly lower trunk and knee flexion, but higher hip flexion at the starting posture. Reduced trunk flexion in older people may be interpreted as a safe lifting strategy since decreased trunk motions are generally associated with reduced low back load (Gregory et al. 2009; Hoozemans et al. 2008) and improved whole-body stability (Puniello et al. 2001). Reduced knee flexion in older participants appeared to be the result of decreased knee joint mobility with aging (James and Parker 1989). It has been suggested that the biological aging of articular structures, such as the reduction on joint cartilage thickness (Hudelmaier et al. 2001), causes the decreased joint motion ranges in the lower limbs of older adults. Larger hip joint flexion in older participants seemed to be a strategy to compensate for the lower trunk and knee flexion so that they were still able to reach the box at the start of lift.

This study also analysed the age-related differences in the ending posture of lift. Older participants had significantly higher shoulder extension and hip flexion in the ending posture compared with younger participants. More extended shoulders in the ending posture would lead to a larger horizontal distance between the load destination and the shoulder joint, and therefore might result in smaller trunk flexion and reduce low back moments as well. Such increased-shoulder-extension strategy was similar as the “stiffening the upper arm” strategy discovered by Chen (2000) in which participants extended their upper arms to avoid trunk overexertion when placing the load onto the shelf. Higher hip flexion in the ending posture in older participants could result in a more forward position of the low back joint, which might decrease low back moment since the horizontal distance between the lifted load and the low back joint was decreased. However, higher hip flexion could give rise to a more forward trunk position, and eventually reduced the whole-body stability for older adults.

The finding regarding the final hip flexion conflicts with Burgess et al. (2009) that reported smaller hip flexion in the ending posture among older adults. This discrepancy might be a

result of differences in lifting frequency between these two studies. In the present study, the lifting frequency was self-selected and typically in the range of 1 to 2 lifts/minute. However, in Burgess et al. (2009), the lifting frequency was controlled to be 20 lifts/minute which is much higher than that in the present study. It was suggested that people would decrease the motion ranges of the lower limbs to avoid the quadriceps muscle fatigue as the lifting frequency increased (Chen 2000; Hagen et al. 1995). Because of the decreased muscle strength with aging (Hurley et al. 1998), older people might induce larger decreases in hip flexion than younger people to prevent the quadriceps muscle fatigue caused by high lifting frequency. Therefore, smaller hip flexion in the final posture was found in Burgess et al. (2009).

In addition to the starting and ending postures, significant age effects were also found in peak trunk and peak knee angular velocities and accelerations. In general, younger participants exhibited higher velocities and accelerations at both joints compared to older ones. Older people performed a less dynamic lifting motion pattern compared to younger people, mainly because they did not have sufficient muscle strength and power to sustain a fast lifting speed (Hurley et al. 1998; Schultz 1995). But, smaller peak trunk and knee angular velocities and accelerations in older people would help decrease low back load and risks of injuries (Bushjoseph et al. 1988; Lindbeck and Arborelius 1991; Marras et al. 1993).

No significant differences were found in low back moment (peak and average) in the lifting phase between younger and older participants, but older participants experienced significantly higher average low back moment in the deposit phase. Hoozemans et al. (2008) found that low back moment was mainly determined by the moment caused by upper body mass if the load mass was no more than 15 kg. They further argued that trunk flexion was a major contributor to low back load since increased trunk flexion resulted in a higher moment caused by upper body mass. As suggested by the existing literature (Flegal et al. 1998;

Hedley et al. 2004; Kuskowska-wolk and Rossner, 1990), older people have higher body mass compared with younger people (i.e. 61.2 kg vs. 54.6 kg on average in this study). So, older people may have higher low back load if they use the same level of trunk flexion as young people do. In the present study, older participants adopted smaller trunk flexion in the lifting phase to offset the effects of their higher body mass. Therefore, low back moment in this phase was similar between two age groups. In the deposit phase, however, older participants had to use similar magnitudes of trunk flexion as younger participants in order to reach the load destination. As a result, older participants experienced higher low back moment in the deposit phase.

Both the peak and cumulative (or time average of) low back loads are contributing factors for LBP (Kumar 1990; McGill 1997; Watersa et al. 2006). In the current study, no significant difference was found in the peak low back moment between two age groups, while older participants showed significantly higher average low back moments than did their younger counterparts. Because of the age-related degeneration at the intervertebral discs (Buckwalter 1995; Roughley 2004), older participants would experience higher risks of LBP than younger ones.

4.4.2 Effects of Destination Height and its Interaction with Age

The main effects of destination height showed that, compared to the elbow height, the wrist height caused higher average low back moment in both of the lifting and deposit phases, and the shoulder height resulted in higher peak low back moment and required larger extensions in upper limbs which might cause fatigue at the upper limbs. Therefore, the elbow destination height should be the optimal height for lifting target in terms of reducing risks of LBP and upper-limb fatigue.

When the destination height was at the shoulder level, the load position was closer to the edge of the reach envelope compared to the other destination height levels, so both younger and older participants had to fully extend their trunk, elbow and shoulder to reach the destination. As a result, between-subject variability in the ending posture was reduced. This can help explain why the age-related differences in the final shoulder extension declined when the destination height increased from the elbow level to shoulder level.

Hoozemans et al. (2008) indicated that low back moment was mainly determined by the moment caused by the upper body mass, and trunk flexion was a major contributor for the low back moment. Because of higher body weights in older participants, the same decrease in trunk flexion would result in larger decreases in the low back moment for older participants compared with younger ones. Increased destination heights resulted in similar decreases in trunk flexion in both age groups, so older participants showed larger decreases in the average low back moment in the deposit phase than younger ones. In the lifting phase, older participants showed significantly decreased average low back moment from wrist to elbow level, while younger participants showed no significant changes in this measure with increased destination height. Therefore, increasing destination height should be a more effective intervention to decrease average low back moment in older adults than in younger adults.

4.4.3 Effects of Load Weight and its Interaction with Age

Increased load weight resulted in decreases in trunk extension velocity and hip joint acceleration. The declined trunk velocity could be a protective motion pattern to offset the increased load weight (Davis and Marras 2000a). With the increased load weight, smaller elbow extension, but larger shoulder extension were observed at the end of lifting. This might be because the strength of the elbow is lower than that of the shoulder, and people have

to depend on larger shoulder extension to deal with heavier load. Increased load weight also gave rise to a more forward body position at both the start (increased hip flexion) and end (increased hip and trunk flexion) of lifting, which would result in more anterior whole-body center of mass, and eventually induced increased whole-body instability.

Older participants did not change final hip flexion across varied load weight levels, while younger participants increased final hip flexion with increased load weights. As discussed earlier (Section 4.4.1 Age effects), larger hip flexion might decrease the low back moment caused by the lifted load by decreasing the horizontal distance between the load and the low back joint. However, larger hip flexion could result in a more forward trunk position, and reduce the whole-body stability. Thus, older adults maintained consistent final hip flexion for maintaining whole-body stability; while younger adults increased final hip flexion in order to compensate for increased low back load due to larger load weights. The final trunk flexion was similar between 15% and 25% MLC load weight levels in older adults, but younger adults showed significant increases in the final trunk flexion from 15% to 25% MLC. Because of the close association between trunk flexion and low back load (Hoozemans et al. 2008), the consistent trunk flexion between 15% and 25% MLC levels among older adults could be an adaptive strategy to avoid substantial increases in low back load.

As mentioned above, low back load was more affected by the trunk posture than the lifted load weight. Since load weight was found to have more substantial adverse effects on trunk kinematics among younger people, we may suspect that increased low back load in the higher load weight condition among younger people was more caused by the change of trunk kinematics than that of load weight. Therefore, in order to reduce low back load among younger people, an effective ergonomic intervention could be instructing people to adopt safe lifting trunk postures (e.g. less flexion). However, trunk kinematics among older people was

not affected by load weight substantially. In order to reduce low back load among older people, decreasing the load weight could be more effective to reduce low back load.

4.4.4 Limitations and Implications

There are limitations in the current study. First, participants were required to keep their feet stationary during the lifting task. Actually, foot movements and footstep strategies could influence trunk kinematics and risks of injury (Delisle et al. 1999; Kollmitzer et al. 2002; Sorensen et al. 2011). Age-related difference in footstep strategies and its effects on lifting biomechanics would be of interest for future research. In addition, only symmetric lifting was studied in the present study. Though symmetric lifting represents a large portion of manual lifting tasks in industry, the load origin and destination are not always located within the sagittal plane in occupational settings. Moreover, asymmetric lifting causes substantial trunk twisting, increased low back loading, and higher risks of LBP (Davis and Marras 2005; Hattori et al. 2000; Marras et al. 1993). To address this limitation, the effects of age and its interactions with work-related factors were studied in asymmetric lifting tasks in Chapter 5.

According to the findings from the present study, older adults tend to use safer lifting strategies (e.g. reduced trunk flexion and lifting speed) compared with younger adults. Thus, increased LBP risks among older adults may not be due to the lifting strategies adopted by them, but because of age-related physiological changes (e.g. decreased strength and larger body mass). Based on these findings and arguments, compared with instructions of safe lifting strategies, physical exercise programs (Sjogren et al. 2006; Moore et al. 2012) aimed at mitigating age-related physiological changes (e.g., improving strength) may be a more effective intervention to reduce LBP risks in lifting for older adults. Actually, a summary of the European guidelines (Burton et al. 2006) for prevention of LBP recommended physical exercise as an important and useful intervention to reduce the risks of LBP. Thus, an

interesting future research direction could be a randomized controlled study to evaluate the effectiveness of different physical exercise programs (e.g., strength improvement training, aerobic exercise, and stretching exercise) in LBP prevention for older adults. Unlike older people, the lifting strategies adopted by younger people are more dangerous in terms of LBP risks. Therefore, instructing younger workers to adopt safer lifting strategies (e.g. reducing trunk flexion) would be effective in LBP risk reduction. In addition, findings from the present study also suggest that optimizing task design, such as setting the destination height at the elbow level, could be effective in reducing the risks of LBP in lifting for both age groups.

Chapter 5 Age-related Biomechanical Differences during Asymmetric Lifting

5.1 Introduction

Asymmetric destination (i.e. destination outside the sagittal plane), which is common in occupational lifting tasks, is one of the work-related factors that could affect lifting biomechanics. In general, asymmetric destination would give rise to greater trunk transverse and lateral motions which significantly affected trunk muscle coactivation (Marras and Mirka 1992; Ning et al. 2011), and increased low back load (Davis and Marras 2005; Marras et al. 1999). Though some studies have been conducted to investigate the effects of asymmetric destination on lifting biomechanics, a limitation with these studies is that most of them examined such effects among younger workers. As discussed in Section 2.2.3, there are various physiological degenerations (e.g., decreased muscle strength, reduced joint flexibility and impaired postural stability) caused by aging process. Therefore, the effects of asymmetric destination on lifting biomechanics in older adults might be different from those in younger ones.

LBP prevalence has been widely reported to be higher in older people (Manchikanti 2000; Hoy et al. 2010). Considering the global aging workforce, there is a need to study age-related biomechanical differences during asymmetric lifting. As reviewed in Section 2.2.3, the only study that examined such age-related differences was conducted by Shin et al. (2006) whose major limitation is that low back load, which is an important contributing factor for LBP in occupations (Coenen et al. 2013; Marras 2006), was not reported.

The main objective of the present study was to examine age-related biomechanical differences during asymmetric lifting. Lifting biomechanics was assessed by trunk kinematics

and low back load. It was hypothesized that there would be significant age-related differences in lifting biomechanics (i.e. trunk kinematics and low back load).

5.2 Methods

The asymmetric lifting trials collected in the lifting experiment (Chapter 3) were analysed in this study to examine the age-related biomechanical differences during asymmetric lifting.

5.2.1 Dependent Measures

The whole lifting period was separated into two phases: lifting phase and deposit phase (Gagnon and Smyth 1992; Hsiang et al. 1998). The lifting phase started from the beginning of lifting and ended when the lifted box was closest to the trunk; the deposit phase began after the lifting phase and continued until the box was deposited at the final destination. These two phases were dependent on the movement of different combinations of joints. In particular, in the lifting phase, participants picked up the box and lifted it close to their trunk, mainly using the extensions of the trunk and lower extremities; in the deposit phase, participants transferred the box to the destination by using upper extremity extension and trunk motions.

5.2.1.1 Trunk Kinematics

Trunk postures and dynamic motions are risk factors for LBP (Davis and Marras 2000a; Hoogendoorn et al. 2000; Punnett et al. 1991). The trunk kinematics measures included 1) the peak sagittal flexion angle, the peak sagittal angular velocity and the peak sagittal angular acceleration, 2) the peak lateral flexion angle, the peak lateral flexion velocity and the peak lateral flexion acceleration, 3) the peak transverse twisting angle, the peak transverse twisting

velocity and the peak transverse twisting acceleration. These measures were examined in the lifting and deposit phases, separately.

To calculate the trunk angles, the local axes of the trunk and pelvis were defined based on the reflective marker positions in Table 3.2. In particular, the Y-axis of the trunk was defined as the vector from L5/S1 to C7/T1, the XY-plane was defined by the L5/S1, C7/T1 and the midpoint of the bilateral shoulder joints, and the Z-axis was perpendicular to the XY-plane and pointing rightward. The right (left) shoulder joint was defined at a point inferiorly to the right (left) acromion, with a fixed distance as 17% of the distance between the bilateral acromions (Rab et al. 2002). For the pelvis, the Z-axis was defined as the vector from left ASIS to right ASIS, the XZ-plane was defined by the midpoint of two PSISs, left ASIS and right ASIS, and the Y-axis was perpendicular to the XZ-plane and pointing upward. The 3-D trunk angles were calculated by decomposing the trunk orientation into Euler angles in the pelvis coordinate system, with the rotation axis order of Z – X – Y (i.e. sagittal flexion – lateral flexion – transverse twisting). The reference angle (0°) for 3-D trunk angles was defined at the standard anatomical position. The positive angles indicated trunk sagittal flexion, lateral rightward flexion and transverse rightward twisting.

5.2.1.2 Low Back Moment

Marras (2006) indicated that low back moment is the most important factor associated with LBP. The 3-D inverse dynamic model (Kingma et al. 1996) discussed in Section 4.2.1 was employed to calculate the low back moment, with inputs of whole-body kinematics, ground reaction forces and segment anthropometry. The moment used as the dependent measures was the magnitude of the resultant moment vector at the low back joint between the trunk and pelvis. Since both peak and accumulative low back loads are risk factors for LBP (Coenen et al. 2013; McGill 1997; Watersa et al. 2006), the peak and average low back total moments

were selected as dependent variables. The peak low back moment was defined over the whole lifting period. The average low back moment was calculated for both phases separately.

5.2.2 Statistical Analysis

Participants conducted three repetitions within each load weight condition, so the dependent measures of these three repetitions was averaged before statistical analysis. Due to the large number of dependent variables, a multivariate ANOVA (MANOVA) was conducted to control the experiment-wise error rate. Then, a mixed ANOVA with repeated measures was performed to evaluate the effects of the independent variables (i.e. age and load weight) and their interactions that showed significant effects in MANOVA. Post-hoc analysis (Bonferroni test) was conducted on the dependent variables significantly affected by load weight. If any significant interaction effect was found, between-subject t-tests were used to test the age effects within each load weight level, and Bonferroni tests were applied to examine the effects of the load weight within each age group. The significance level was set at 0.05 for all analyses. Prior to the analysis, the assumptions of ANOVA (i.e. independence of observations, normality and homogeneous variance of residuals) were assessed using the approaches suggested by Montgomery (2005). A violation of the homogeneous variance assumption was detected, and it was remedied with a natural log transformation. The sphericity assumption of the repeated measures ANOVA was tested using Mauchly's sphericity test, and Greenhouse-Geisser adjustment was applied to remedy violations.

5.3 Results

As in Chapter 4, the postural index (PI) defined by Burgess and Abernethy (1997) was also used to examine the lifting strategy adopted by participants in this chapter. Burgess and Abernethy (1997) suggested that full stoop and squat strategies corresponded to $PI \leq 0.11$ and

≥ 0.8 , respectively. The average values of the PI were 0.58 (SD: 0.09) in younger adults and 0.49 (SD: 0.10) in older adults. Therefore, both age groups chose a lifting strategy intermediate between full stoop and full squat.

The result of the MANOVA showed significant effects of age and load weight (i.e. p -values < 0.05), although no significant interaction effect of the two factors was found. The ANOVA results also showed significant effects of age and load weight on trunk kinematics and low back load. Specifically, older participants showed significantly decreased peak trunk sagittal flexion in both the lifting and deposit phases and smaller peak trunk transverse twisting angles in the deposit phase (Table 5.1). The peak trunk angular velocities and accelerations in all three planes (i.e. sagittal, lateral and transverse) in both the lifting and deposit phases were significantly smaller in older participants compared with younger ones. No significant age-related differences were observed in the peak low back moment and the average low back moment in the lifting phase. However, older participants showed significantly higher average low back moment in the deposit phase.

As expected, increased load weight resulted in significant increases in all the dependent variables about low back moments (Table 5.1). In the lifting phase, the peak trunk lateral flexion velocity and acceleration as well as the peak trunk transverse twisting acceleration significantly increased when the load weight increased from 5% MLC to 15% MLC. In the deposit phase, significant increases in the peak trunk sagittal flexion angle, lateral flexion acceleration and transverse twisting angle were found from 5% to 25% MLC load weight level.

Table 5.1 The results of the mixed ANOVA

	Age			Load Weight			
	Young Mean (SD)	Older Mean (SD)	<i>p</i> -value	5% MLC Mean (SD)	15% MLC Mean (SD)	25% MLC Mean (SD)	<i>p</i> -value
<i>Peak Trunk Kinematics (Lifting Phase)</i>							
Sagittal flexion angle (°)	45.28 (6.55)	30.74 (5.84)	<0.001	37.82 (9.80)	37.51 (9.59)	37.76 (9.71)	0.535
Sagittal angular velocity (°/s)	61.57 (14.17)	33.96 (9.37)	<0.001	47.96 (17.57)	47.77 (18.93)	45.75 (18.93)	0.160
Sagittal angular acceleration (°/s ²)	173.77 (49.63)	99.42 (51.62)	0.002	137.39 (58.07)	136.32 (69.82)	131.24 (62.67)	0.475
Lateral flexion angle (°)	1.88 (2.73)	0.90 (2.70)	0.403	1.14 (2.60)	1.59 (2.95)	1.38 (2.77)	0.108
Lateral flexion velocity (°/s)	11.98 (4.14)	7.85 (2.93)	0.007	8.93 (3.32) ^A	10.58 (4.83) ^B	9.96 (4.03) ^{A,B}	0.004
Lateral flexion acceleration (°/s ²)	83.89 (33.15)	55.02 (18.70)	0.003	59.38 (20.53) ^A	77.66 (39.56) ^B	69.44 (25.40) ^B	0.001
Transverse twisting angle (°)	6.99 (3.65)	5.47 (5.00)	0.421	6.01 (4.33)	6.25 (4.32)	6.31 (4.83)	0.615
Transverse twisting velocity (°/s)	30.90 (9.63)	20.56 (9.15)	0.011	23.60 (10.18)	26.95 (11.13)	25.98 (10.87)	0.051
Transverse twisting acceleration (°/s ²)	174.60 (59.75)	123.55 (39.68)	0.009	129.31 (37.86) ^A	158.31 (62.01) ^B	156.27 (62.54) ^B	0.003
<i>Peak Trunk Kinematics (Deposit Phase)</i>							
Sagittal flexion angle (°)	19.30 (3.31)	12.13 (4.88)	<0.001	14.59 (5.70) ^A	15.73 (5.45) ^{A,B}	16.36 (5.45) ^B	0.016
Sagittal angular velocity (°/s)	48.23 (14.82)	23.85 (7.31)	<0.001	35.17 (16.53)	36.15 (17.97)	35.22 (16.52)	0.980
Sagittal angular acceleration (°/s ²)	188.17 (55.90)	103.13 (39.85)	<0.001	142.17 (62.36)	143.93 (67.72)	145.30 (65.20)	0.700
Lateral flexion angle (°)	1.70 (2.09)	0.50 (2.43)	0.218	1.29 (2.20)	1.11 (2.39)	0.82 (2.50)	0.137
Lateral flexion velocity (°/s)	10.17 (2.82)	6.44 (3.09)	0.002	7.85 (3.47)	8.02 (3.16)	8.79 (3.88)	0.218
Lateral flexion acceleration (°/s ²)	69.73 (28.85)	39.63 (15.38)	<0.001	48.95 (24.98) ^A	53.40 (28.45) ^{A,B}	59.72 (28.31) ^B	0.007
Transverse twisting angle (°)	21.35 (4.07)	16.64 (3.29)	0.004	17.91 (5.06) ^A	19.22 (3.78) ^{A,B}	19.56 (4.16) ^B	0.003
Transverse twisting velocity (°/s)	40.54 (11.83)	26.10 (7.94)	0.002	32.08 (11.67)	32.85 (12.79)	34.09 (12.86)	0.188
Transverse twisting acceleration (°/s ²)	191.50 (84.02)	114.92 (39.64)	0.007	148.72 (78.10)	150.31 (79.09)	155.62 (70.53)	0.753
<i>Low Back Moment</i>							
Peak moment (Nm)	161.98(32.34)	159.27(39.63)	0.856	149.12(32.62) ^A	161.07(34.31) ^B	171.51(38.95) ^C	<0.001
Average moment – Lifting phase (Nm)	117.59(26.93)	121.41(31.11)	0.736	105.51(22.79) ^A	120.01(27.92) ^B	133.21(30.09) ^C	<0.001
Average moment – Deposit phase (Nm)	68.15(13.37)	89.77(25.29)	0.005	66.75(17.07) ^A	79.14(19.32) ^B	92.41(25.20) ^C	<0.001

* The conditions with the same superscripted letter (A, B or C) were not significantly different from each other.

5.4 Discussion

The main purpose of this study was to examine the age effects on lifting biomechanics during asymmetric lifting. Both younger and older participants were instructed to conduct asymmetric lifting with different load weights. The results showed significant age-related differences on lifting biomechanics which was assessed using trunk kinematics and low back moments.

5.4.1 Age Effects

5.4.1.1 Trunk Kinematics

Older participants showed significantly smaller peak trunk sagittal flexion in the lifting phase than did younger participants in asymmetric lifting (Table 5.1). This result conflicts with Shin et al. (2006) who reported no age-related differences in trunk flexion during asymmetric lifting. This discrepancy might be due to the different experimental settings between these two studies. In the present study, the load was transferred away from the sagittal plane; while in Shin et al. (2006), the load was transferred towards the sagittal plane.

Trunk flexion is closely associated with low back load, and has been regarded as a risk factor for occupational LBP (Punnett et al. 1991; Ozguler et al. 2000; Wickström and Pentti 1998). Hoozemans et al. (2008) found that low back moment during lifting was largely determined by the moment of the upper body mass. Increased trunk flexion would generate a larger moment arm for the upper body weight about the low back joint, and eventually increase the low back moment. Furthermore, by combining vivo kinematics measurements and model studies, Bazrgari et al. (2007) found significantly increased back muscle forces and spinal compression/shear forces with higher trunk flexion.

One may argue that the decreased trunk flexion in older adults could be associated with age-related decreases in trunk flexibility and range of motion. However, the peak trunk flexion

angles of the older adults observed in this study (30.74°) were much smaller than the maximum trunk flexion angle (around 50°) and well within the range of motion of adults in their sixties (Troke et al. 2001, 2005). In other words, older participants have a large margin in their flexibility/range of motion capacity when conducting the lifting tasks in the present experiment. Therefore, a reduction in trunk flexibility and range of motion with age may not be the major cause of the age-related decrease in trunk flexion. Instead, changes of trunk kinematics could mainly reflect the adoption of a safer lifting strategy to reduce low back load by older adults.

Although the higher trunk flexion could increase low back load, this posture pattern might provide benefits in terms of energy saving for younger participants. Maduri, Pearson and Wilson (2008) indicated that the trunk extensor muscles and associated tendons store elastic energy when they are stretched during trunk flexion, and release the energy as they are shortened in trunk extension. It was argued that this stretching-shortening pattern could reduce the energy expenditure when raising the torso in lifting. Therefore, younger participants in the present study might take advantage of the stretching-shortening behavior in trunk muscles and tendons to save the energy at the cost of increasing their low back loads. Older adults, however, gave the first priority to reducing the low back load by keeping a more erect torso positions, probably because of the age-related degenerations in trunk muscle and intervertebral discs (Adams et al. 2002; Doherty 2003; Roughley 2004).

The trunk lateral flexion was much smaller than the trunk transverse twisting (Table 5.1), which indicated that participants mainly applied trunk twisting instead of trunk lateral flexion to transfer the load to the asymmetric destination. Older participants showed significantly lower trunk transverse twisting than did younger ones in the deposit phase, which could be a protective lifting strategy to reduce LBP risks for older adults. Trunk twisting has been identified as a risk factor for occupational LBP (Hoogendoorn et al. 2000; Hoy et al. 2005;

Meksawia 2012; Punnett et al. 1991). As indicated in Kumar (2004), spinal rotation significantly reduced trunk strength (up to 75% compared with spinal extension) and increased trunk muscle loadings (up to 300%). Also, the spinal compression forces during lifting are markedly increased with combined trunk flexion and twisting because greater muscle recruitments are involved to stabilize the torso in asymmetric postures (Granata and Wilson 2001).

During asymmetric lifting, trunk angular velocities and accelerations in all the three planes (i.e. the sagittal, lateral and transverse planes) within both lifting and deposit phases were smaller in older participants compared with their younger counterparts (Table 5.1). High trunk angular velocities have been found to be a risk factor for occupational LBP (Marras et al. 1995; Norman et al. 1998). Davis and Marras (2000a) indicated that increases of trunk angular velocities in any of the three planes (i.e., sagittal, lateral and transverse) would severely reduce the dynamic trunk muscle strength. They further pointed out that because of the decreased trunk muscle strength, greater muscle forces were needed to offset external loads, and eventually resulted in increased muscle activities and spinal loads. Based on these arguments, self-selected reduced trunk angular velocities in older adults could be a protective strategy to compensate their decreased trunk strength and prevent high spinal loads during asymmetric lifting. As for younger adults, although the higher trunk angular velocities and accelerations could induce larger spinal loads, these loads might still be in the safe margin for younger people because of their higher trunk muscle strengths (Singh et al. 2013). Since increased lifting velocities reduced the total lifting duration, the high trunk dynamics could be an efficient lifting strategy for younger adults at the expense of increases in low back loadings.

5.4.1.2 Low back moment

Hoozemans et al. (2008) argued that trunk flexion was a major contributor to low back load. Also, as discussed previously, increased trunk angular velocities and accelerations would result in greater low back loads. Therefore, according to the reduced trunk flexion and less dynamic trunk motions within the lifting phase in older participants, it might be expected that older people should have shown significantly reduced low back moment compared to their younger counterparts in this phase. However, in contrast to this expectation, we did not find any significant difference in the peak and average low back moments within the lifting phase between two age groups. This might be resulted from the higher body mass in older adults compared with that in younger ones (i.e. 61.2 kg vs. 54.6 kg on average in this study). Actually, increased body mass is one of the physiological changes with aging process (Flegal et al. 1998; Hedley et al. 2004; Kuskowska-wolk and Rossner, 1990). The higher body mass of older adults resulted in increases in low back moment by enlarging the moment of the upper body mass, and therefore counteracted the reductions in low back moment caused by the decreased trunk flexion and its velocities and accelerations.

The increased body mass with aging should also account for the significantly higher average low back moment during the deposit phase in older adults. Hoozemans et al. (2008) indicated that low back moment was largely determined by the moment of the upper body mass. Although older adults showed significantly smaller peak trunk flexion than did younger adults in the deposit phase, the difference in the trunk flexion between two age groups in this phase was relatively smaller than that in the lifting phase (7.17° in the deposit phase vs. 14.53° in the lifting phase). Therefore, the higher body mass in older adults increased the moment of the upper body mass, which eventually resulted in significantly higher low back moment in older adults than that in younger ones during the deposit phase.

Both peak and cumulative low back loads are contributing factors for LBP (Coenen et al. 2013; McGill 1997; Watersa et al. 2006), and the low back moment is closely associated with occupational low back disorders (Marras 2006). In the current study, no significant age-related difference was found in peak low back moment. However, older adults showed significantly higher average low back moment, which would lead to higher cumulative low back loadings compared with younger adults. Considering muscle strength decrease and intervertebral disc degeneration with aging (Doherty 2003; Roughley 2004), older adults would have significantly increased risks of LBP than younger ones.

5.4.2 Load Weight Effects

Higher load weight resulted in increased peak trunk sagittal flexion and transverse twisting angles in the deposit phase (Table 5.1). As discussed in Section 5.4.1, increased trunk sagittal flexion is associated with higher low back load, and higher trunk twisting is a risk factor for LBP since it would reduce trunk muscle strength and increase low back load. Increased load weight also induced higher peak trunk lateral flexion velocity and acceleration as well as peak trunk transverse twisting acceleration in the lifting phase. The increases in these trunk dynamics measures were associated with higher low back load as well. Therefore, given that higher load weight itself would increase low back load substantially, such load-weight-related trunk kinematics changes would further increase risks of LBP in higher load weight condition during asymmetric lifting. In Chapter 4, it was found that participants decreased trunk sagittal extension velocities with higher load weight during symmetric lifting so as to offset the increases in load weight, which could be interpreted as a protective lifting strategy. The differences between symmetric and asymmetric lifting in the load weight effects on trunk kinematics may be one cause of relative greater LBP risks in asymmetric condition compared with symmetric condition, especially when it comes to high load weight.

5.4.3 Interactions between Age and Load Weight

Older adults generally have lower muscle strength and load lifting capacities compared with younger adults. Existing literatures showed that muscle strength would significantly affect lifting posture and motion patterns (Chen 2000; Sasaki et al. 2008; Li and Zhang 2009; Zhang and Buhr 2002). Therefore, one may expect different biomechanical responses to increasing load weight between younger and older age groups during asymmetric lifting. However, no significant interaction effect between age and load weight was discovered in the current study. This might be caused by the load weight selection protocol in this study. In particular, the load weight was standardized to each individual's lifting capacity (5%, 15% and 25% MLC) instead of using absolute values. Thus, younger and older adults were exposed to the same level of challenge at each load weight level in terms of their individual muscle strength.

5.4.4 Limitations and Practical Implications

Some limitations of the present study should be noted. First, foot movement was controlled in the experiment. Existing literature showed that changes in footstep strategies and base of support could affect lifting kinematics, low back load as well as whole-body stability (Delisle et al. 1999; Kollmitzer et al. 2002; Sorensen et al. 2011). Therefore, age-related differences in footstep strategy during lifting and the corresponding effects on lifting biomechanics would be of interest in the future research. Second, only low back moment was examined, and spinal compression and shear forces were not used to assess low back load in the present study. Spinal compression and shear forces are considered as the direct indicators for low back loading, and have been widely used to evaluate the risk of low back injuries in previous literatures (Davis and Marras 2005; Faber et al. 2008; Hoozemans et al. 2008; Kingma et al. 2004). However, Marras (2006) indicated that the external low back moment is the most important contributing factor to occupational LBP. In addition, van Dieën and Kingma (2005) found that low back moment was highly correlated with spinal total and compression forces.

Therefore, the low back moment examined in this study is an effective measurement to assess risks of LBP during lifting.

The findings of the present study suggest that, compared with younger adults, older adults adopt safer lifting strategies (i.e. reduced trunk sagittal flexion and transverse twisting, decreased trunk velocities) in order to reduce low back load during asymmetric lifting. However, they might still have higher risks of LBP than younger ones because of the larger cumulative low back load and the age-related reduced trunk muscle strength. Therefore, the higher risks of low back injury during asymmetric lifting in older people might not be caused by their posture and motion patterns, but mainly resulted from the age-related physiological changes (e.g. decreased muscle strength and increased body weight). Based on these findings and arguments, compared with instructions of safe asymmetric lifting techniques, physical exercise programs for alleviating the physiological degenerations with aging (e.g. improving trunk muscle strength) may be a more effective intervention in reducing LBP incidence for older adults. Actually, Burton (2005) suggested that physical exercise is one of the most promising approaches to reduce the incidence and recurrence of LBP. The lifting strategies of younger adults are more efficient in terms of reducing energy costs and whole lifting time duration, though such strategies are more likely to induce low back injuries compared with older people. Therefore, instructions of safe asymmetric lifting techniques might be effective to reduce LBP risks for younger adults.

Chapter 6 Age Effects on Postural Stability in Lifting Tasks

6.1 Introduction

Falls are a leading cause of various occupational injuries and fatalities (Dong et al. 2013; Lin et al. 2008; Yoon and Lockhart 2006). In the US, The annual direct cost of occupational injuries caused by falls is estimated to exceed \$43.8 billion by 2020 (Englander et al. 1996). A substantial proportion of occupational falls results from loss of postural stability. Load lifting and transferring makes perturbations to postural stability (Kollmitzer et al. 2002). In fact, fall incidence has been reported to be particularly high in industries requiring manual material handling (Bureau of Labor and Statistics 2009). Postural stability in load lifting and transferring has been examined in existing literature which suggested large load weights (Catena et al. 2010), absence of load knowledge (van der Burg et al. 2000), and a slippery surface (Catena et al. 2011) as risk factors for falling.

Older adults are associated with poorer postural stability and higher fall risks, probably because of age-related physiological changes, including impaired sensory systems, decreased muscle strength and longer reaction time (Sturnieks et al. 2008). Given the worldwide aging workforce, there is a need to examine age effects on postural stability during load lifting. As discussed in Section 2.2.3, the only study that has investigated age effects on postural stability during lifting was done by Shin et al. (2006). A limitation with Shin et al. (2006) is that they did not report measurements of the whole-body center of mass (COM) which provides insight into dynamic postural control mechanisms (Hof et al., 2005).

The main purpose of this study was to investigate the effects of age on postural stability during lifting. Postural stability was assessed by the inclination angles between the COM and COP as well as ground reaction forces. It was hypothesized that there would be significant

age-related differences in postural stability during lifting. Specifically, given higher risks of falling among older adults, older adults were expected to have larger COM-COP inclination angles and larger peak horizontal ground reaction forces than younger adults.

6.2 Methods

Both the symmetric and asymmetric lifting trials with load destinations at the waist height level were examined in this study to investigate the effects of age, load weight and task asymmetry on postural stability during lifting.

6.2.1 Dependent Measures

Postural stability was assessed using the peak horizontal ground reaction forces and peak COM-COP (centre of mass – centre of pressure) inclination angles (Lee and Chou 2007; Shin et al. 2006). The horizontal ground reaction forces were defined as the peak forces during the whole lifting period in four directions: anterior, posterior, leftward and rightward. The COM was defined as the centre of mass for the combination of the body and the lifted load. The body COM was computed as the weighted sum of each body segment's center-of-mass from the 12-segment model including the head, trunk, upper arms, lower arms, thighs, shanks, and feet. The parameters for estimating body segments' center-of-mass locations were provided by der Leva (1996). The load COM was determined as the center of the box. The COP was defined as the net COP of the two feet measured by the force plates. Specifically, the net COP in either the A/P or M/L directions was calculated as Eq. 6.1 (Winter 1995):

$$COP_{net} = COP_l \frac{R_{vl}}{R_{vl} + R_{vr}} + COP_r \frac{R_{vr}}{R_{vl} + R_{vr}} \quad (\text{Eq. 6.1})$$

in which COP_l and COP_r are the COP under left and right foot, respectively; R_{vl} and R_{vr} are the ground reaction forces under left and right foot, respectively. To reduce the error in estimating the COM-COP inclination angles, the COP position measured by the force plates

was modified using an optimization method proposed by Heiss and Pagnacco (2002). The COM-COP inclination angles were calculated in the anterior-posterior (A/P) and medial-lateral (M/L) directions, respectively. In particular, the A/P COM-COP inclination angle was defined as the angle from a vertical line through the COP to the projection of the line connecting the COM and COP onto the sagittal plane. The M/L COM-COP inclination angle was defined as the angle from the vertical line through the COP to the projection of the COM-COP-connecting line onto the frontal plane.

6.2.2 Statistical Analysis

The dependent measures were averaged over the three repetitions in each task condition. A three-way mixed ANOVA with repeated measures was performed to assess the effects of age, load weight and task asymmetry as well as their interactions. In the post-hoc analysis, Bonferroni test was conducted on the dependent measures that were significantly affected by load weight. If a significant interaction between age and one of the work-related factors (i.e. task asymmetry and load weight) was found, between-subject *t*-tests were employed within each work-related factor level to assess the age effects, Bonferroni tests were applied to examine the load weight effects within each age group, and repeated-measures *t*-test was employed to evaluate the effects of the task asymmetry in each age group. The significance level was set at 0.05 for all analysis. Before all the analyses, the assumptions of ANOVA (i.e. independence of observations, normality and homogeneous variance of residuals) were tested using the approaches suggested by Montgomery (2005). A violation of the homogeneous variance assumption was detected, and it was remedied with a natural log transformation. The sphericity assumption of the repeated measures ANOVA was assessed using Mauchly's sphericity test, and Greenhouse-Geisser adjustment was applied to remedy violations.

6.3 Results

The statistical analysis showed significant age effects on peak horizontal ground reaction forces (Table 6.1). Specifically, older adults showed significantly smaller peak posterior and leftward ground reaction forces than did younger adults. No significant age-related differences in peak COM-COP inclination angles were found. Besides, there were no significant interaction effects between age and work-related factors (i.e. load weight and task asymmetry) on any selected dependent variables.

The results showed significant effects of task asymmetry on all dependent variables (Table 6.2). The peak A/P COM-COP inclination angle was significantly smaller, and the peak M/L COM-COP inclination angle was significantly larger in asymmetric lifting than those in symmetric lifting. The peak anterior force was significantly larger in symmetric lifting, while the peak posterior, leftward and rightward forces were significantly larger in asymmetric lifting. The increased load weight resulted in significant increases in the peak M/L COM-COP inclination angle, the anterior and rightward horizontal ground reaction forces.

Table 6.1 Age main effects and the interaction effects between age and work-related factors

	Age			Age × Task Asymmetry	Age × load Weight
	Young Mean (SD)	Older Mean (SD)	p-value	p-value	p-value
<i>Peak COM-COP Inclination Angles</i>					
Anterior-posterior angle (°)	3.42 (0.73)	3.08 (1.04)	0.216	0.723	0.754
medial-lateral angle (°)	2.69 (1.09)	2.46 (1.12)	0.236	0.620	0.974
<i>Peak Horizontal Ground Reaction Forces</i>					
Anterior force (N)	30.12 (14.51)	24.19 (14.78)	0.223	0.522	0.488
Posterior force (N)	32.52 (18.02)	14.14 (16.71)	0.009	0.091	0.282
Leftward force (N)	27.60 (19.68)	16.96 (13.20)	0.028	0.130	0.230
Rightward force (N)	18.27 (13.37)	15.82 (14.34)	0.502	0.182	0.164

Table 6.2 Main effects of load weight and task asymmetry

	Task Asymmetry			Load Weight			
	Symmetry Mean (SD)	Asymmetry Mean (SD)	p-value	5% MLC Mean (SD)	15% MLC Mean (SD)	25% MLC Mean (SD)	p-value
<i>Peak COM-COP Inclination Angles</i>							
Anterior-posterior angle (°)	3.50 (1.01)	2.98 (0.73)	0.004	3.27 (0.92)	3.09 (0.89)	3.36 (0.95)	0.184
medial-lateral angle (°)	1.76 (0.50)	3.39 (0.90)	<0.001	2.46 (1.07) ^A	2.58 (1.08) ^{A,B}	2.68 (1.15) ^B	0.046
<i>Peak Horizontal Ground Reaction Forces</i>							
Anterior force (N)	32.90 (15.28)	21.15 (11.99)	<0.001	23.17 (15.02) ^A	26.64 (13.20) ^B	31.26 (15.56) ^C	<0.001
Posterior force (N)	20.48 (21.01)	25.38 (17.87)	0.01	21.34 (17.78)	21.19 (16.46)	26.25 (23.75)	0.070
Leftward force (N)	12.17 (10.61)	31.94 (17.29)	<0.001	17.22 (12.98) ^A	22.00 (16.81) ^B	26.93 (20.54) ^C	<0.001
Rightward force (N)	7.92 (8.06)	26.28 (12.26)	<0.001	16.12 (12.52)	16.91 (13.84)	18.27 (15.26)	0.260

* The conditions with the same superscripted letter (A, B or C) were not significantly different from each other

6.4 Discussion

The main purpose of this study was to examine age-related differences in postural stability during lifting. Both younger and older participants were instructed to conduct symmetric and asymmetric lifting tasks with different load weights in a controlled experiment. Postural stability was assessed using the peak horizontal ground reaction forces and peak COM-COP inclination angles.

The results showed significantly larger peak posterior and leftward horizontal ground reaction forces in younger adults than those in older ones (Table 6.1). This could be resulted from the higher whole body lifting velocities and accelerations in younger adults (please see Chapter 4 and Chapter 5). With higher lifting velocities and accelerations, larger external forces were needed to decelerate the motions of the human body and the lifted load so as to deposit the load at the destination.

Higher horizontal ground reaction forces were associated with increased slip incidences (Cham and Redfern 2002; Lockhart et al. 2003). Therefore, the reduced lifting velocities in older adults which could lead to smaller horizontal ground reaction forces could be interpreted as a protective strategy to enhance their postural stability. However, such a protective strategy might be only effective in the condition when the lifting frequency is low or self-selected. As shown in Shin et al. (2006), there were no age-related differences in the ground reaction forces when the lifting frequency was controlled at around 8 lifts/minute. In our work, the lifting frequency was self-selected and generally ranged from 1 to 2 lifts/minute. Based on this argument, self-selected or low lifting frequency should be adopted in lifting task designs for older adults so that they can decrease their lifting velocities to enhance their balance.

The results showed no significant differences in the COM-COP inclination angles between two age groups (Table 6.1). This might result from the load weight selection protocol in this study. Specifically, the load weight was normalized by individual lifting capacity (i.e., 5%, 15% and 25% MLC). Older adults generally have smaller MLC than did younger adults (Table 3.1). Thus, the lifted load weights were lower in older participants than those in younger ones. This suggested that the postural perturbations due to the lifted load were smaller in older adults in our experimental study. Higher load weight was reported to be associated with increased risks of loss of balance during load transferring tasks (Catena et al. 2010). In the present study, increased load weight also resulted in increases in the M/L COM-COP inclination angle, the peak anterior and leftward horizontal ground reaction forces (Table 6.2). Therefore, postural stability would be worse in older adults if they lifted the same absolute load weights as younger adults did, and the postural perturbations due to the same load weights would result in higher fall risks in older adults. Thus, in order to maintain postural stability in older adults, the lifted load weight should be adjusted according to individual strengths.

The results showed significantly increases in the M/L COM-COM inclination angle, the leftward and rightward horizontal ground reaction forces in the asymmetric lifting compared with symmetric lifting. Therefore, asymmetric lifting brought about increased risks of loss of balance in the frontal plane. The effects of task asymmetry on postural stability in the sagittal plane were mixed. Compared to the symmetric condition, asymmetric lifting brought about significant decreases in the A/P COM-COP inclination angle and the anterior horizontal ground reaction forces, but significant increases in the posterior horizontal ground reaction force. Winter (1995) indicated that the human body was more vulnerable to a fall in the frontal plane compared with that in the sagittal plane, since the body generally has better postural control capabilities in the sagittal plane than in the frontal plane. Thus, the

asymmetric condition might induce higher risks of falls compared with the symmetric condition, because of decreased postural stability in the frontal plane during asymmetric lifting.

Older adults are generally associated with decreased muscle strength and deteriorated postural stability compared with younger ones (Doherty 2003; Sturnieks et al. 2008). Therefore, one may expect that these age-related physiological changes would bring about different postural control responses to increased load weight and lifting asymmetry. However, no significant interaction effects between age and the work-related factors (i.e., load weight and task asymmetry) were observed. This might be resulted from the selected lifting protocols in the current study. Specifically, the load weight was adjusted according to each individual's lifting capacity (5%, 15% and 25% MLC), so both age groups were exposed to the same challenge level caused by the load weight in terms of their individual muscle strength. The asymmetric angle of the destination was only 60°, which might not be large enough to induce significant interaction effects, compared with those angles (e.g., 90° and 135°) used in previous studies (Allread et al. 1996) on task asymmetry.

There are limitations in the present study. First, the foot movements were restricted during the lifting tasks. Actually, footstep strategies could affect postural control and stability during lifting (Delisle et al. 1999; Kollmitzer et al. 2002). Thus, Age-related difference in footstep strategies and its effects on postural stability would be of interest for future research. In addition, the lifting conditions in this study (e.g., normal ground surfaces and low load weights adjusted to individuals' lifting capacities) might not be challenging enough to induce high risks of falls. Therefore, age effects on postural control will be examined in more challenging lifting task conditions (e.g., higher load weights and slippery ground surfaces with low friction coefficients) in future studies.

In summary, we found that when performing lifting tasks, older adults adopted a more protective postural control strategy (i.e. reduce lifting velocities) to enhance postural stability. In addition, both higher load weight and task asymmetry are associated with poorer postural stability and higher fall risks. Based on the findings from this study, it might be suggested that in order to maintain postural stability during lifting, especially in older adults, the lifted load weight be adjusted according to individual strengths, and asymmetric lifting be avoided if possible.

Chapter 7 Lifting Motion Simulation Using a Hybrid Approach

7.1 Introduction

Human motion simulation is an effective ergonomics tool in analysing and improving the designs of products, workplaces and occupational manual tasks (Chaffin 2005). The human musculoskeletal system is mechanically redundant which means there are always more degrees of freedom (DOFs) than those necessary to perform the motions during manual tasks. Therefore, motion simulation is an ill-posed problem which involves no unique solutions. The major challenge of the motion simulation is to identify the realistic ones among all possible motions which are able to fulfil the required manual task demands.

To solve this ill-posed problem, two groups of methods, including data-based and optimization-based, have been developed. Both of the methods have distinct advantages and limitations which have been discussed in Section 2.3.3. To combine the advantages of both data-based and optimization-based methods, hybrid approaches have been developed in recent research (Pasciuto et al. 2014; Xiang et al. 2012b). Pasciuto et al. (2014), for instance, simulated clutch pedal depression motions by combining a knowledge-based and a data-based objective function. The knowledge-based objective function was defined based on a minimal energy criterion, while the data-based one was defined as the minimum difference in the velocity profiles between the simulated and actual motions. Another example was given by Xiang et al. (2012b) who developed a hybrid predictive dynamics (HPD) model for lifting motion simulation. The objective of the optimization procedure in this model was to minimize the weighted sum of a knowledge-based (i.e., minimal dynamic effort) and a data-based (i.e., minimal error between the simulated and actual motions) objective functions by using a multi-objective optimization (MOO) approach. The main drawback of the current

hybrid approaches was that there were no effective ways in determining the appropriate weightings of each objective function in the MOO. Xiang et al. (2012b) arbitrarily set the weighting coefficients of the knowledge- and data-based objective functions at 0.3 and 0.7, respectively. Pasciuto et al. (2014) compared multiple pairs of possible weightings and selected the one associated with the minimal error between the simulated and actual motions. Apparently, this comparison procedure was inefficient and time-consuming.

The objective of this study was to develop a novel hybrid model for lifting motion simulation. Instead of using the data-based objective function, the actual motions were used to construct constraints in the optimization procedures. This can help avoid the difficulty in determining the effective weightings for the MOO (Pasciuto et al. 2014; Xiang et al. 2012b). Specifically, lifting motions performed by both young and older adults under varied task conditions were simulated using the hybrid model. The model performance was evaluated by comparing the simulated motions with their corresponding actual ones.

7.2 Lifting Data Collection

Actual lifting motion data are needed for model development and evaluation. Many lifting tasks in occupations are symmetric (i.e., in the sagittal plane), so this simulation model focuses on predicting two-dimensional (2-D) lifting motions. Therefore, the data used in this study were from the symmetric lifting motions collected in the experiment as described in Chapter 3. These collected motions were performed by 23 subjects (11 younger and 12 older). The task scenarios were defined by the three load weights (5%, 15% and 25% of each individual's maximum lifting capacity) and three destination heights (the wrist, elbow and shoulder levels during the stance posture). The lifting task in each scenario was repeated three times by each subject.

The motion data of one older subject and 42 lifting trials from other 11 subjects are excluded because of the significant asymmetric movement patterns. The remaining data are divided into two sets for the purposes of model development and evaluation, respectively. Specifically, the lifting motions from 12 subjects are selected to formulate a database for the model development. These 12 subjects consist of 6 younger (age: 23.3 (SD: 2.1), height: 168.2cm (SD: 8.5cm), weight: 52.3kg (SD: 4.7kg)) and 6 older (age: 64.5 (SD: 5.2), height: 163.7cm (SD: 7.7cm), weight: 61.3kg (SD: 11.5kg)), and each age group contains 3 males and 3 females. The lifting motions performed by the other 10 subjects (5 younger and 5 older) are used for model evaluation. Both the younger (age: 24 (SD: 2.1), height: 165.6cm (SD: 5.0cm), weight: 57.4kg (SD: 9.1kg)) and older (age: 68.6 (SD: 6.2), height: 159.6cm (SD: 7.8cm), weight: 58kg (SD: 8.2kg)) age groups contain three males and two females. There are totally 315 and 237 lifting trials for model development and evaluation, respectively.

7.3 Hybrid Lifting Motion Simulation Model

The inputs of the model are the lifting task scenario and the performer's attributes. The task scenario is defined by the lifted load weight as well as the initial and final load positions. The performer's attributes include age, gender, body weight and segment lengths. Similar with previous studies (Chang et al. 2001; Hsiang and Ayoub 1994; Lin et al. 1999; Qu and Nussbaum 2009), the starting and ending body postures of the simulated motion are specified by the corresponding actual motion data. The total lifting time duration are also specified before simulation. The outputs of the model are the predicted motions represented as the joint angular trajectories (angle-time functions). The whole model consists of four components: a lifting motion database, a physical human body model, an optimization model, and a joint angular trajectory model. The model structure and the simulation process are illustrated in Fig.

7.1. The database consists of the collected motion data for model development as described in Section 7.2. The details of the other components and the simulation process are discussed as follows.

Figure 7.1 The structure of the simulation model and the simulation process

7.3.1 Physical Human Body Model

A 2-D linkage model was developed to represent the human body during sagittally symmetric lifting (Fig 7.2). The global coordinate system was also shown in Fig 7.2, with its origin at

the ankle joint centre. The model consists of five segments: forearm, upper arm, trunk, thigh and shank ($L_1 \sim L_5$). The mass of the head and neck are incorporated in the trunk segment, and these three body parts are combined as one rigid link. The inertia parameters of each segment (i.e., the mass, the moment of inertia and the centre of mass (COM) position on the segment) are determined according to de Leva (1996). Six joints connecting these segments are wrist, elbow, shoulder, hip, knee and ankle ($J_0 \sim J_5$). The five joints from the elbow to ankle ($J_1 \sim J_5$) are modelled as revolute joints with one degree of freedom. Therefore, given the segment lengths, the whole body configuration can be specified with the five joint angles ($\theta_i, i = 1, \dots, 5$) which are defined as the angles of the segments respect to the horizontal axis X .

Figure 7.2 The Human Linkage Model

Given the lifted load weight, the segment lengths and inertia parameters, the joint reaction forces and moments can be calculated based on the whole body kinematics. Specifically, the calculation involves analysing each body segment in a top-down sequence (forearm – upper arm – trunk – thigh – shank) using inverse dynamics. Fig 7.3 shows a diagram of one

segment L_i , and its joint reaction forces and moments are calculated by the following dynamic equations of motion (Hibbeler et al. 2009):

$$\begin{aligned} \mathbf{F}_i - \mathbf{F}_{i-1} + m_i \mathbf{g} &= m_i \mathbf{a}_i \\ M_i - M_{i-1} - \mathbf{r}_{G_i, J_{i-1}} \times \mathbf{F}_{i-1} + \mathbf{r}_{G_i, J_i} \times \mathbf{F}_i &= I_i \ddot{\theta}_i \end{aligned} \quad (\text{Eq. 7.1})$$

in which $\mathbf{F}_i$ and M_i are the force vector and moment about the i -th joint (J_i) exerted on the i -th segment (L_i), respectively; $\mathbf{r}_{G_i, J_{i-1}}$ and $\mathbf{r}_{G_i, J_i}$ are the vectors pointing from the COM (G_i) of the segment L_i to the joint J_{i-1} and J_i , respectively; $\mathbf{g}$ is the gravity vector; m_i and I_i are the mass and moment of inertia of the segment L_i , respectively; $\mathbf{a}_i$ is the linear acceleration of the COM G_i ; $\ddot{\theta}_i$ is the angular acceleration of the segment L_i ; and $\times$ denotes a vector product.

Figure 7.3 The diagram of a segment with its joint reaction forces and moments

7.3.2 Optimization Model

Lifting motion prediction is finally formulated into a non-linear optimization problem. In the optimization procedures, the joint angular trajectories are predicted by minimizing an objective function, subject to various constraints (Eq. 7.2):

$$\begin{array}{llll}
\text{Find} & \theta_i(t) & i = 1, \dots, 5 \\
\text{Minimize} & Obj(\theta_i) & & \\
\text{Subject to} & Con_j^{Eq}(\theta_i) = 0 & j = 1, \dots, m & \text{(Eq. 7.2)} \\
& Con_k^{Ineq}(\theta_i) \leq 0 & k = 1, \dots, n &
\end{array}$$

in which $\theta_i(t)$ are the joint angular trajectories, Obj is the objective function, Con_j^{Eq} and Con_k^{Ineq} are equality and inequality constraints, respectively.

7.3.2.1 Objective Function

The objective function is minimizing the joint strength utilization (JSU) (Hsiang and McGorry 1997) which is defined as the time integral of the square sum of the ratios between the joint moments and the corresponding joint strengths (Eq. 7.3):

$$Obj = \int_0^T \sum_{i=1}^5 \left(\frac{M_i(t)}{S_i} \right)^2 dt \quad \text{(Eq. 7.3)}$$

in which M_i is the moment of joint i , S_i is the strength of joint i , and T is the total lifting time duration. The JSU was found to be associated with the physical efforts spent in the lifting task, and has been extensively used for lifting motion prediction in previous studies (Hsiang and Ayoub 1994; Hsiang and McGorry 1997; Lin et al. 1999).

Given the segmental lengths and inertia properties, the joint moments $M_i(t)$ can be calculated from the joint angular trajectories $\theta_i(t)$ using Eq. 7.1. The joint strengths S_i were defined as the 50% percentile of the static joint moment strengths of the male and female populations, respectively, according to Chaffin et al. (2006).

7.3.2.2 Constraints

The constraints of the optimization consist of joint angle limits, data-based joint angular velocity constraints, joint moment limits, postural balance constraints, and body-load and shelf-load collision avoidance.

1. Joint angle limits

The joint angles defined by the orientations of their adjacent segments (referred to as included joint angles) are illustrated as α_i ($i = 1, \dots, 5$) in Fig. 7.4. These included joint angles α_i can be derived from the joint angles θ_i using the equations in Table 7.1. Their motion ranges (Table 7.1) were obtained from Chaffin et al. (2006).

Figure 7.4 Definition of the included joint angles

Table 7.1 Motion ranges of the included joint angles ($^{\circ}$)

Included joint angles	Lower bound (α_i^L)	Upper bound (α_i^U)
Elbow: $\alpha_1 = 180 - \theta_1 + \theta_2$	38	180
Shoulder: $\alpha_2 = 180 - \theta_3 + \theta_2$	-61	188
Hip: $\alpha_3 = 180 - \theta_4 + \theta_3$	67	180
Knee: $\alpha_4 = 180 - \theta_4 + \theta_5$	67	180
Ankle: $\alpha_5 = \theta_5$	55	128

The included joint angle limits are defined by their motion ranges as follows:

$$\alpha_i^L \leq \alpha_i(t) \leq \alpha_i^U \quad i=1,...,5, \quad t \in [0, T] \quad (\text{Eq. 7.4})$$

where α_i^L and α_i^U are the lower and upper boundaries of the i -th included joint angle, and T is the total time duration.

2. Data-based joint angular velocity constraints

In this hybrid model, the time derivatives of the included joint angular trajectories (i.e., the included joint angular velocities) of the predicted motions were assumed to be similar with the motions in the lifting motion database. Specifically, the time functions of the included joint angular velocities ($\dot{\alpha}_i(t), i=1,...,5$) of the predicted motions are bounded by a set of functional constraints defined by the lifting motion database.

To define these data-based constraints, the data in the lifting motion database are first analysed using a functional regression technique (Faraway 1997). The response variables of the regression are the normalized time functions of the included joint angular velocities $\dot{\alpha}_i(\tau)$, where τ is the normalized time with $\tau = 0$ and $\tau = 1$ corresponding to the start and the end of the motion, respectively. The experimental results in Chapter 4 revealed significant effects of load weight, destination height (i.e., shelf height) and age on the joint angular velocities, but no significant interactions among them. Therefore, the predictors in the regression are defined by load weight, destination height, and age. A quadratic regression model is developed as follows (Eq. 7.5):

$$\dot{\alpha}_i^M(\tau) = \beta_i^0(\tau) + \beta_i^W(\tau)W + \beta_i^H(\tau)H + \beta_i^A(\tau)A + \beta_i^{W^2}(\tau)W^2 + \beta_i^{H^2}(\tau)H^2 + \varepsilon_i(\tau) \quad (\text{Eq. 7.5})$$

$$i = 1, ..., 5; \quad \tau \in [0, 1]$$

where $\dot{\alpha}_i^M(\tau)$ is the response variable predicted by the regression, which denotes the mean of the i -th included joint angular velocity function of the normalized time ; W , H and A are the predictors: W is the load weight defined as the percentage of the subject's maximum lifting capacity, H is the destination height calculated as the percentage of the body height, A is a binary accounting for the subject's age: 0 = younger adults and 1 = older adults; β_i^0 , β_i^W , β_i^H , β_i^A , $\beta_i^{W^2}$, $\beta_i^{H^2}$, ε_i are the estimated time functions from the regression analysis, and $\varepsilon_i(\tau)$ is the estimated standard deviation of $\dot{\alpha}_i^M(\tau)$.

Given the total lifting duration T , the functions with normalized time τ ($\dot{\alpha}_i^M(\tau), \varepsilon_i(\tau)$) are transformed to the functions with absolute time t ($\dot{\alpha}_i^M(t), \varepsilon_i(t)$) by replacing τ with t/T . The constraints on the included joint angular velocities are defined as follows (Eq. 7.6):

$$\dot{\alpha}_i^M(t) - 1.96\varepsilon_i(t) \leq \dot{\alpha}_i(t) \leq \dot{\alpha}_i^M(t) + 1.96\varepsilon_i(t) \quad t \in [0, T] \quad (\text{Eq. 7.6})$$

where $\dot{\alpha}_i(t)$ is the predicted included angular velocity of the i -th joint at the time point t , and $\left[\dot{\alpha}_i^M(t) - 1.96\varepsilon_i(t), \dot{\alpha}_i^M(t) + 1.96\varepsilon_i(t) \right]$ defines the 95% confidence interval .

3. Joint moment limits

It is assumed that the joint moments would not exceed their corresponding strengths during the whole lifting period. Therefore, the constraints on the joint moments are defined as follows:

$$M_i^2(t) - S_i^2 \leq 0 \quad i = 1, \dots, 5, \quad t \in [0, T] \quad (\text{Eq. 7.7})$$

where M_i and S_i are the joint moment and static strength of the i -th joint in Eq. 7.3, respectively

4. Postural stability constraints

To maintain the postural stability, the horizontal COM position about the body-load system should be located within the foot supporting area. Therefore, the postural stability constraints can be defined as follows:

$$x_{Heel} \leq x_{COM}(t) \leq x_{Toe} \quad t \in [0, T] \quad (\text{Eq. 7.8})$$

where x_{COM} is the horizontal position of the centre of mass for the combination of the body and the lifted load. The body COM is computed as the weighted sum of each body segment's COM. The parameters for estimating body segments' COM locations are derived from der Leva (1996). The load COM is determined as the centre of the box. x_{Heel} and x_{Toe} are defined as the horizontal positions of the corresponding reflective markers (i.e., heel and 5th metatarsal in Table 3.2), respectively.

5. Body-load collision avoidance

The collision between the human body (trunk, thigh and shank) and the lifted load should be avoided during lifting. The lifted load in this study is a square box (width $\times$ height: 0.24m $\times$ 0.26m). For simplicity, the box dimension is approximated to be a circle with its centre at the box geometric centre and its diameter as the box height (Circle A in Fig 7.5). The trunk, thigh and shank dimensions are modelled as three rectangles (the dashed lines in Fig 7.5) with their segment lines at the centre. The widths of these rectangles (i.e., the trunk, thigh and shank dimensions) are estimated from the measured reflective marker positions in Chapter 3. Specifically, the trunk dimension is calculated as the distance from the ASIS to the PSIS in the sagittal plane; the thigh and shank dimensions are approximated as the distance between the lateral and medial epicondyles of femur. In this way, the constraints of the body-load collision avoidance are defined based on the distances between the box centre and the trunk, thigh and shank segment lines:

$$\|\mathbf{P}_{Body}(t) - \mathbf{P}_{BC}(t)\| \geq \frac{1}{2}(H_{Box} + Dim_{Seg}) \quad t \in [0, T] \quad (\text{Eq. 7.9})$$

where $\mathbf{P}_{Body}$ is the position of any point on the trunk, thigh or shank segment lines; $\mathbf{P}_{BC}$ is the position of the box centre, $\|\cdot\|$ denotes the Euclidean distance between $\mathbf{P}_{Body}$ and $\mathbf{P}_{BC}$; H_{Box} is the box height; and Dim_{Seg} represents the segment (i.e., trunk, thigh or shank) dimensions.

Figure 7.5 Definitions of the body-load and shelf-load collision avoidance constraints

6. Shelf-load collision avoidance

To define the constraint of the shelf-load collision avoidance, an imaginary circle is placed right below the shelf (Circle B in Fig 7.5), and the box should bypass this circle during lifting. The circle was tangent to the shelf, and its diameter is set as the same as the box height. The horizontal distance from the circle centre to the shelf edge is the radius of circle. The lifted box is approximated in the same way as in the body-load collision avoidance constraints. Thus, the constraints of the shelf-load collision avoidance can be defined by calculating the distance from the box centre to the imaginary circle centre:

$$\|\mathbf{P}_{BC}(t) - \mathbf{P}_{CC}\| \geq \frac{1}{2}(H_{Box} + D_{Circle}) \quad t \in [0, T] \quad (\text{Eq. 7.10})$$

where $\mathbf{P}_{BC}$ is the position of the box centre, $\mathbf{P}_{CC}$ is the position of the imaginary circle centre, H_{Box} is the box height, and D_{Circle} is the diameter of the imaginary circle.

7.3.3 Joint Angular Trajectory Model

Human motions are composed of infinite postures changing with time. Therefore, the optimization problem in Eq. 7.2 is actually of infinite dimension. To solve this infinite-dimension problem, the joint angular trajectories are represented using polynomial functions, and the parameters in these functions become the design variables in the optimization procedure. Specifically, each joint angular trajectory $\theta_i(t)$ is represented by a Hermite polynomial (Kincaid and Cheney 2002) interpolated by the joint angles at certain time points during lifting:

$$\begin{aligned} \theta_i(t) &= f(\theta_{i0}, \theta_{iT}, \theta_i(t_l), T, t) & i = 1, \dots, 5 \\ \dot{\theta}_{i0} &= \dot{\theta}_{iT} = 0, \quad \ddot{\theta}_{iT} = 0 \end{aligned} \quad (\text{Eq. 7.11})$$

in which θ_{i0} and θ_{iT} are the joint angles in the starting and ending postures, respectively; t_l represents the time points at 20%, 40%, 60% and 80% of the total time duration T ; $\dot{\theta}_{i0}$ and $\dot{\theta}_{iT}$ are the initial and final joint angular velocities which are assumed to be zero; and the final joint angular acceleration $\ddot{\theta}_{iT}$ is assumed to be zero (Hsiang and Ayoub 1994).

Therefore, the motion prediction problem in Eq. 7.2 is transformed into an optimization problem in which the design variables, defined by a set of joint angles $\theta_i(t_l)$ and denoted as a vector $\mathbf{\theta}$, are predicted:

$$\begin{aligned}
&\text{Find} && \boldsymbol{\theta} \\
&\text{Minimize} && \text{Obj}(\boldsymbol{\theta}) \\
&\text{Subject to} && \text{Con}_j^{\text{Eq}}(\boldsymbol{\theta}) = 0 \quad j = 1, \dots, m \\
& && \text{Con}_k^{\text{Ineq}}(\boldsymbol{\theta}) \leq 0 \quad k = 1, \dots, n \\
&\boldsymbol{\theta} = [\theta_1(t_1), \theta_1(t_2), \dots, \theta_i(t_l), \dots, \theta_5(t_4)] && i = 1, \dots, 5 \\
&t_l = 0.2T, 0.4T, 0.6T, 0.8T && l = 1, \dots, 4
\end{aligned} \tag{Eq. 7.12}$$

7.3.4 Simulation Process

The whole simulation process is illustrated in Fig. 7.1. Details of each step in the process are as follows:

Step 1. The functional regression equations (Eq. 7.5) are first specified using the actual motions in the database. Given the simulated task scenario (i.e., load weight and destination height), the participant's age and the total lifting duration, the parameters $\dot{\alpha}_i^M(t)$ are determined using Eq. 7.5, and further used to construct the data-based joint angular velocity constraints for the optimization model.

Step 2. The initial values of the design variable vector ($\boldsymbol{\theta}_0$) are determined using actual motions selected from the motion database. It should be noted that the selected actual motions must be consistent with the simulated one in age attribute (younger or older) and task scenario (i.e., load weight and destination height). Multiple motion trials could be retrieved from the database for simulating one lift. If this is the case, these trials are used to simulate the lifting motions of interest separately. The simulated motion that generates the minimum value of the objective function (Eq. 7.3) is selected as the final prediction result.

Step 3. Once the initial values of the design variable vector ($\boldsymbol{\theta}_0$) and the data-based constraints ($\dot{\alpha}_i^M(t), \varepsilon_i(t)$) are determined, the lifting motions are predicted by solving the optimization problem (Eq. 7.13) using an iterative searching method illustrated in the dashed-

line square in Fig. 7.1. The iteration starts with the initial values of the design variable vector θ_0 , the initial and final body postures and the total time duration as the inputs. Each searching iteration consists 3 sub-steps:

- 1) The joint angular trajectories $\theta_i(t)$ are first formulated based on the inputs using the joint angular trajectory model.
- 2) The joint angular trajectories $\theta_i(t)$ are then sent to the physical human body model to calculate various kinematic and kinetic variables, including the joint moments, the included joint angles and velocities, the box position and the whole body configurations throughout the total lifting duration.
- 3) These variables are used to evaluate the objective function and constraints in the optimization model. The optimization model generates intermediate design variable vector θ_{Inter} after each searching iteration. θ_{Inter} is used as the input to the joint angular trajectory model for the next iteration.

Step 4. Finally, the optimal design variable vector θ_{Optim} are predicted by minimizing the objective function subject to all the constraints.

The optimization routine was conducted using the sequential quadratic programming algorithm (SQP) implemented by the optimization toolbox in MATLAB.

7.4 Motion Simulation and Model Evaluation

The 237 motions of 10 subjects mentioned in Section 7.2 were predicted and used to evaluate the model performance. Specifically, the task scenario, the personal attributes, the starting and ending postures and the total time duration of each actual motion was input into the

model to generate a predicted motion. To evaluate the model, the predicted motion was compared with its corresponding actual motion. The divergence between these two was quantified using the mean, median and root-mean-squared (RMS) absolute errors of the joint angular trajectories. The mean square errors (MSE) have been used to evaluate the prediction accuracy in previous research (Hsiang and Ayoub 1994; Lin et al. 1999), and these values were also calculated for each joint angle in this study. The equations for calculating the mean, median and RMS errors of the overall motions and the MSE for the j th joint angle (MSE_j) are as follows (Eq. 7.13 - 16):

$$Mean\ Error = \frac{\sum_{j=1}^5 \sum_{i=1}^n \left| \hat{\theta}_{ij} - \theta_{ij} \right|}{5n} \quad (Eq. 7.13)$$

$$Median\ Error = Median\ value\ among\ \frac{\sum_{j=1}^5 \left| \hat{\theta}_{ij} - \theta_{ij} \right|}{5} \quad (i=1,...,n) \quad (Eq. 7.14)$$

$$RMS = \sqrt{\frac{\sum_{j=1}^5 \sum_{i=1}^n \left(\hat{\theta}_{ij} - \theta_{ij} \right)^2}{5n}} \quad (Eq. 7.15)$$

$$MSE_j = \frac{\sum_{i=1}^n \left(\hat{\theta}_{ij} - \theta_{ij} \right)^2}{5n} \quad (Eq. 7.16)$$

in which $\hat{\theta}_{ij}$ and θ_{ij} are the predicted and actual the j th joint angle values at time frame i , respectively; n is the total time frame number. As suggested by Qu and Nussbaum (2009), the correlation coefficients between predicted and actual joint angular trajectories were also used to quantify the correspondence between the two motions.

7.5 Simulation Results

Summary results regarding the mean, median and RMS absolute errors between actual and predicted lifting motions were presented in Table 7.2. The mean and median errors across varied task scenarios and age groups were less than 10° . Overall, the errors were higher in the older age group than the younger. Table 7.3 showed the MSE measures across joints. In particular, the predicted elbow and shoulder motions had much larger errors than the hip, knee and ankle motions did. The correlation coefficients for all the joints were larger than 0.8 (Table 7.4), which indicated high correspondence between the predicted and actual motions. Comparisons between the actual and predicted motions for two representative lifting trials were illustrated in Fig. 7.6. These two trials were performed by one younger and one older subject, respectively, with the load weight of 15% MLC and the shelf height at the elbow level.

Figure 7.6 The actual and simulated joint angular trajectories for one younger and one older subject, with the load weight of 15% maximum lifting capacity (MLC) and shelf height at elbow level

Table 7.2 Mean, median and RMS absolute errors (°) between actual and predicted motions

	The number of trials	Mean	Median	RMS
<i>Age</i>				
Younger	122	6.69	6.63	10.26
Older	115	8.41	9.01	13.02
<i>Load weight</i>				
5% MLC	76	7.37	7.54	11.23
15% MLC	84	7.55	7.85	11.72
25% MLC	77	7.66	7.95	11.82
<i>Shelf height</i>				
Wrist	83	7.69	8.02	11.98
Elbow	83	7.10	7.57	10.75
Shoulder	71	7.84	7.75	12.13
<i>Overall</i>	237	7.53	7.78	11.60

Table 7.3 Mean square errors (square degree) across joints

	Elbow	Shoulder	Hip	Knee	Ankle
<i>Age</i>					
Younger	235.05	224.94	49.45	35.38	24.61
Older	428.33	332.07	65.30	46.29	37.75
<i>Load weight</i>					
5% MLC	270.58	289.28	46.86	42.42	31.73
15% MLC	324.26	280.49	62.08	44.67	31.04
25% MLC	391.32	260.82	61.90	34.59	30.21
<i>Shelf height</i>					
Wrist	450.36	173.72	78.15	37.91	21.22
Elbow	229.29	269.08	52.27	38.15	26.51
Shoulder	303.13	406.73	38.28	46.86	47.64
<i>Overall</i>	328.83	276.92	57.14	40.67	30.99

Table 7.4 Correlation coefficients between actual and predicted joint angular trajectories

	Elbow	Shoulder	Hip	Knee	Ankle
<i>Age</i>					
Younger	0.94	0.95	0.96	0.98	0.90
Older	0.87	0.93	0.96	0.96	0.72
<i>Load weight</i>					
5% MLC	0.92	0.94	0.97	0.98	0.82
15% MLC	0.91	0.94	0.96	0.97	0.79
25% MLC	0.89	0.94	0.96	0.98	0.83
<i>Shelf height</i>					
Wrist	0.81	0.94	0.95	0.97	0.89
Elbow	0.96	0.94	0.97	0.98	0.85
Shoulder	0.96	0.95	0.98	0.97	0.67
<i>Overall</i>	0.91	0.94	0.96	0.97	0.81

7.6 Discussion

The main purpose of this study was to develop and evaluate a hybrid model for symmetric lifting motion simulation. A 2-D five-segment body model was used, and the lifting motions performed by younger and older individuals under varied task conditions (i.e., different load weights and destination heights) were predicted by solving an optimization problem. Specifically, the optimization objective was to minimize the joint strength utilization (JSU) subject to various constraints which included the data-based constraints extracted from the actual motions.

When evaluating human motion simulation models, 10° has been suggested as the limiting criterion value for the absolute mean joint angle errors (Beck and Chaffin 1992; Chaffin and Page 1994). From this perspective, the current model was able to accurately simulate 2-D lifting motions, since the mean and median joint angle errors were less than 10° (Tables 7.2). The simulation results were comparable with those of previous lifting motion simulation models. Hsiang and Ayoub (1994) developed an optimization-based model to simulate lifting motions, using the same objective function (i.e., JSU) as in the present study. The mean MSEs in the current study (Tables 7.3) were substantially smaller than those in Hsiang and Ayoub (1994) across all the joints. Perez and Nussbaum (2008) constructed a neural network model to predict lifting motions, and the RMS errors for each joint were less than 20° . In the present study, the average RMS errors for a combination of all joints are 11.60° (Table 7.2). When considering each joint separately, the RMS error is generally less than 20° (Table 7.3). Lin et al. (1999) developed a dynamic model for predicting 2-D lifting motions. The mean MSEs in their study were larger for the hip and knee joints (hip: 78.4; knee: 148.6; unit: square degree), while smaller for the other joints (elbow: 105.0; shoulder 270.2; ankle: 20.9; unit: square degree) than those in the present study (Table 7.3).

The error measures (i.e., mean, median and RMS errors) in the present study were larger than those in Qu and Nussbaum (2009) who simulated lifting motions using a fuzzy-logic-based model. However, compared with those in the study of Qu and Nussbaum (2009), the correlation coefficients in the present study (Tables 7.4) were closer to 1, which may suggest higher fidelity of the simulated motions in the present study. In Qu and Nussbaum (2009), the motions used for model development (i.e., developing the fuzzy rule) and evaluation were from the same participants. In other words, their model cannot account for inter-individual variances. However, the motion data for model development and evaluation in the present study are from different groups of participants. Given that the lifting strategies across individuals might be substantially different, the prediction errors of the fuzzy-logic-based model may increase significantly when it is used for between-subject motion simulation.

The absolute joint angle errors were larger for the elbow and shoulder than for the hip, knee and ankle (Tables 7.3), which was also found in previous studies (Chang et al. 2001; Lin et al. 1999; Qu and Nussbaum 2009). This might be resulted from the substantially larger motion ranges of the elbow and shoulder (mean: 142° and 241°) compared with those of the hip, knee and ankle (mean: 113° , 113° and 73°) (Table 7.1). Actually, a larger joint motion range is expected to lead to increased prediction errors.

There were larger joint angle errors in older participants (Tables 7.2 and 7.3). This might result from the constraint definitions in the optimization component of this model. Specifically, the data used for the joint angle and moment limits were determined based on younger subjects (Chaffin et al. 2006), and the age-related decreases in the joint ranges and muscle strengths were not considered in the definitions of the corresponding constraints (i.e., joint angle and moment limits). Thus, to better simulate lifting motions for older adults in future research, age-related physiological and biomechanical changes (e.g., decreased muscle strength and joint mobility) should be taken into account when defining the constraints in the

optimization component in future research. Another possible explanation for the larger errors of older adults may be that the minimal-physical-effort is not the only performance criterion used by older subjects to control their lifting motions. For instance, given the impaired postural stability in older adults (Sturnieks et al. 2008), enhancing body balance could be another performance criterion. Therefore, as suggested by Chang et al. (2001), multiple performance criteria might be necessary in order to predict more accurate lifting motions for older adults. In addition, the functional regression model in the data-based joint angular velocity constraints is constructed based on both younger and older adults' actual motions. This could also be a cause of the larger simulation error in older adults, because the younger adults' motion data may affect the regression results for older adults. Therefore, developing two separated regression models for younger and older adults based on their own corresponding actual motions, respectively, may have a positive effect in improving the prediction accuracy of the simulation model.

The proposed methodology for lifting motion simulation is not restricted to specific database. In other words, practitioners could develop their own database when using the proposed methodology to predict lifting motions and assess the risks of manual lifting tasks. However, it should be noted that some adjustments to the model might be necessary to accommodate the new lifting motion database. In particular, the current database only includes the task conditions defined by two factors (i.e., load weight and destination height). If additional task factors (e.g., coupling) are considered in the new database, more predictors accounting for the additional task factors should be added into the functional regression model in the data-based joint angular velocity constraints (Eq. 7.5), and a more generalized motion selection rules (e.g., Park et al. 2008) should be adopted for determining the initial design variable vector (Section 7.3.4).

As reviewed in Section 2.3.3, there are two major limitations in the existing human motion simulation models: 1) the performance criteria in the optimization-based models are difficult to validate; 2) the effects of inter-individual differences (i.e. age differences) on simulated motions are not taken into account. The use of a hybrid approach for lifting motion simulation in this chapter can help address the two limitations.

First, it incorporated the actual motion information in an optimization-based framework. Specifically, the joint angular velocities were constrained by the time-dependent boundaries extracted from the actual motions, so the shape of the joint angular trajectories resembled that of the actual ones. Thus, compared with the optimization-based models, this model assured the validity of the predicted motions. Second, the boundaries varied based on different task scenarios and performer's attributes (i.e., load weight, destination height and age), so the effects of these task and individual factors on the movement patterns were taken into account in the simulation. Besides, compared with the existing hybrid simulation model (Pasciuto et al. 2014; Xiang et al. 2012b), the present model used the actual motions to construct the data-based constraints instead of including a data-based objective function in the optimization. In this way, multi-objective optimization (MOO) was avoided in the present study so that the whole procedures were simplified compared to the previous studies.

There are several limitations in this model. First, only 2-D lifting motions were simulated. Asymmetric lifting is common in occupations, and often associated with higher risks of LBP (Davis and Marras 2005). Therefore, the presented model should be applied to simulate 3-D lifting motions in the next step. Second, the motion data used for model development and evaluation are performed in the same task scenarios but by different participants. Therefore, the simulation results can reflect the proposed model's capability of predicting motions between subjects in the same task scenarios. However, to validate the capacity of the model on simulating lifting motions between subjects in novel task scenarios, further model

evaluation using motions of different scenarios and participants should be conducted in future studies. Third, the present study only used one performance criterion (i.e., minimal physical effort). As discussed before, the larger joint angle errors in older adults might be attributed to this limitation. In Chapter 8, using multiple performance criteria in lifting motion simulation will be further investigated.

In summary, a novel hybrid model was developed for 2-D lifting motion simulation in this study. The lifting motions were simulated by solving an optimization problem. In the optimization procedure, the objective function was developed based on the performance criterion of minimal effort, and data-based constraints were imposed on the joint angular velocities of the simulated motions. By using the data-based motion constraints, this model ensured the validity of the simulated motions, and was able to account for the effects of work-related factors (i.e., destination height and load weight) and personal attributes (i.e., age). Lifting tasks performed by younger and older adults under varied task conditions were simulated. Comparisons were made between the simulated and actual motions. The comparisons indicated that this model is comparable with existing motion simulation models in terms of the prediction accuracy.

Chapter 8 Lifting Motion Simulation Using a Hybrid Multi-objective Optimization Approach

8.1 Introduction

In Chapter 7, a hybrid model was developed for lifting motion simulation. In this model, data-based constraints were incorporated into an optimization framework, which enhanced the accuracy and validity of the predicted motions. However, the absolute joint angle errors for the older subjects were relative larger than those for the younger ones. These findings might be because only one performance criterion (i.e., minimal physical effort) was applied in the optimization procedures. As suggested by Chang et al. (2001), more than one performance criterion might be needed to better predict and explain the lifting behaviour.

As reviewed in Section 2.3.1.2, Xiang et al. (2010) simulated lifting motions by using a multi-objective optimization (MOO) approach in which two performance measures (i.e., dynamic effort and stability) were optimized simultaneously. Compared with using the dynamic effort as a single performance measure, there were no significant improvements on the prediction accuracy after aggregating the two performance measures with the MOO approach. Thus, the maximum stability may not be an effective performance criterion for lifting motion simulation, and there is a need to further investigate alternative performance criteria which can be used in the MOO approach for better lifting motion simulation.

Another limitation of the existing models for lifting motion simulation is that they were only used to predict motions for young and/or middle-aged (20 - 40 years) adults (Chang et al. 2001; Dysart and Woldstad 1996; Hsiang and Mcgorry 1997; Lin et al. 1999; Xiang et al. 2010). Lifting motions of older adults (> 55 years) have not been extensively studied using the motion simulation method. In fact, experimental studies in Chapter 4 and 5 showed

significant distinctions in lifting motion patterns between young and older adults, which implies that there exist age-related differences in the mechanisms for planning lifting motions.

To address the limitations of the existing lifting motion simulation models, the objectives of the present study are twofold. First, we aimed to propose a novel lifting motion simulation model by incorporating the MOO approach into the hybrid model proposed in Chapter 7. The second objective was to investigate age-related differences in the mechanisms for lifting motion planning using the proposed lifting simulation model. Specifically, two performance criteria, including minimum physical effort and maximum smoothness of the external load motion, were investigated. The objective function in the MOO was defined as the weighted sum of the performance measures derived from these two criteria. Lifting motions of younger (20 – 30 years) and older (> 55 years) adults were simulated separately by using different weighting parameters in the MOO approach. The optimal weighting parameters of the two age groups were specified by those that could lead to the most accurate lifting motion prediction results.

8.2 Lifting Motion Simulation Model

The motion prediction model in this study is the same as that in Chapter 7 (Fig. 7.1) except the objective function in the optimization component. Instead of using a single-objective optimization, the weighted-sum MOO approach (Marler and Arora 2004) is applied to simulate lifting motions. Specifically, a composite objective function (Obj_{Com}) is constructed based on two single objective functions (Obj_{s1} and Obj_{s2}), and then minimized to predict optimal motions. Therefore, the optimization problem in Eq. 7.12 is transformed as follows:

$$\begin{aligned}
& \text{Find} && \boldsymbol{\theta} \\
& \text{Minimize} && Obj_{Com} = F(\boldsymbol{\theta}, Obj_{s1}, Obj_{s2}) \\
& \text{Subject to} && Con_j^{Eq}(\boldsymbol{\theta}) = 0 \quad j = 1, \dots, m \\
& && Con_k^{Ineq}(\boldsymbol{\theta}) \leq 0 \quad k = 1, \dots, n \\
& && \boldsymbol{\theta} = [\theta_1(t_1), \theta_1(t_2), \dots, \theta_i(t_l), \dots, \theta_5(t_4)], \quad i = 1, \dots, 5 \\
& && t_l = 0.2T, 0.4T, 0.6T, 0.8T, \quad l = 1, \dots, 4
\end{aligned} \tag{Eq. 8.1}$$

in which $\boldsymbol{\theta}$ is the vector of the design variables defined as the joint angles θ_i at time t_l ; T is the total lifting time duration; $F(\bullet)$ indicates that Obj_{Com} is a function of the design variables $\boldsymbol{\theta}$ and the single objective functions (Obj_{s1} and Obj_{s2}). The definitions of the single objective functions and the formulation of the composite objective function are discussed in details as follows.

8.2.1 Single Objective Functions

Two objective functions are defined corresponding to two performance criteria: minimum physical effort and maximum smoothness of the load motion. The objective function of minimum physical effort (Obj_{Effort}) is defined as Eq. 7.3 in Chapter 7. The second objective function is defined based on the performance criterion of maximum load motion smoothness (Obj_{Smooth}), as this criterion is associated with reduced low back load which is a critical issue in lifting motion control (Hsiang and McGorry 1997). The objective function Obj_{Smooth} is defined as minimizing the jerk (i.e., maximizing the smoothness) of the end-effector (i.e. the hands and lifted load) motion. The mathematical equation of this objective function is:

$$Obj_{Smooth} = \int_0^T \left(\left(\frac{d^3 x_L}{dt^3} \right)^2 + \left(\frac{d^3 y_L}{dt^3} \right)^2 \right) dt \tag{Eq. 8.2}$$

in which x_L and y_L are the horizontal and vertical coordinates of the load geometric centre, respectively.

8.2.2 Formulation of the Composite Objective Function

The composite objective function Obj_{Com} is defined as a weighted sum of the single objective functions Obj_{Effort} and Obj_{Smooth} after normalization:

$$\begin{aligned} Obj_{Com} &= w_E N(Obj_{Effort}) + w_S N(Obj_{Smooth}) \\ w_E + w_S &= 1 \end{aligned} \quad (\text{Eq. 8.3})$$

where w_E and w_S are the weightings for Obj_{Effort} and Obj_{Smooth} , respectively, with their ranges from 0 to 1. $N(\bullet)$ is a normalization operator defined by Marler and Arora (2005). Specifically, the normalization of a single objective function $N(Obj_{Sp})$ ($p=1$ or 2) is defined as follows:

$$N(Obj_{Sp}) = \frac{Obj_{Sp}(\theta) - Obj_{Sp}^{Min}}{Obj_{Sp}^{Max} - Obj_{Sp}^{Min}} \quad (\text{Eq. 8.4})$$

in which Obj_{Sp}^{Min} is the minimum value of Obj_{Sp} , and Obj_{Sp}^{Max} is defined as $Obj_{Sp}(\theta^*)$ where θ^* is the point that minimizes the other single objective function Obj_{Sq} ($q=1, 2; q \neq p$).

8.3 Motion Simulation and Model Evaluation

The same lifting trials used for the model evaluation in Chapter 7 were simulated in this study. The simulation process in this study is the same as that in Chapter 7 (See Section 7.3.4 and Figure 7.1), except that the objective function in the current model is the composite objective function defined in Section 8.2.2. Besides, the values of the weighting parameters for the MOO approach are also specified as an input before simulation. To examine the effects of the weightings on the prediction results, a series of weighting values were evaluated in the simulation. Specifically, the weightings were ranged from 0 to 1 with an equal interval as 0.1:

$$w_E = 0:0.1:1, \quad w_S = 1 - w_E \quad (\text{Eq. 8.5})$$

The model performance was evaluated by comparing the simulated motions with their corresponding actual ones. The discrepancy between these two was quantified using the root-mean-squared (RMS) absolute joint angle errors. The errors of the predicted motions based on different weightings were compared within younger and older groups, respectively, so as to determine the optimal weightings for each age group. Finally, the mean, median and RMS absolute errors of the whole lifting motions, the MSEs and correlation coefficients between each predicted and actual joint angular trajectory were calculated as in Section 7.4 when utilizing the optimal weightings of young and older groups.

8.4 Simulation Results

Figure 8.1 shows the overall RMS absolute joint angle errors between the simulated and actual motions for different weighting values in the younger and older subjects. The weightings resulting in minimum errors for younger and older age groups are $w_E = 0.5/w_S = 0.5$ and $w_E = 0.1/w_S = 0.9$, respectively. The mean, median and RMS absolute errors of the whole lifting motions, the MSEs and correlation coefficients between each predicted and actual joint angular trajectory when utilizing the optimal weightings are presented in Tables 8.1-3. When using Obj_{Effort} and Obj_{Smooth} as the single objective functions, the overall RMS absolute joint angle errors in the older adults are 13.02° and 10.62° , respectively. After using the MOO approach, the error for the older subjects decreases to 9.33° when $w_E = 0.1$ and $w_S = 0.9$ (Fig. 8.1).

For the younger subjects, the minimum overall RMS error across different weightings is 9.61° when w_E and w_S are both 0.5 (Fig. 8.1). The errors are 10.26° and 12.17° when only using Obj_{Effort} (i.e., $w_E = 1.0$ and $w_S = 0$) and Obj_{Jerk} (i.e., $w_E = 0$ and $w_S = 1.0$) in the single objective optimization, respectively.

Figure 8.1 RMS absolute joint angle errors (°) under different weighting values for the younger and older subjects

Table 8.1 Mean, median and RMS absolute errors (°) between actual and predicted motions

	The number of trials	Mean	Median	RMS
<i>Age</i>				
Younger	122	6.28	6.26	9.61
Older	115	6.09	6.37	9.33
<i>Load weight</i>				
5% MLC	76	5.98	6.21	9.01
15% MLC	84	6.12	6.21	9.46
25% MLC	77	6.48	6.54	9.95
<i>Shelf height</i>				
Wrist	83	6.31	6.57	9.75
Elbow	83	5.77	5.85	8.75
Shoulder	71	6.54	6.57	10.02
<i>Overall</i>	237	6.19	6.32	9.48

Table 8.2 Mean square errors (square degree) across joints

	Elbow	Shoulder	Hip	Knee	Ankle
<i>Age</i>					
Younger	188.24	201.22	51.58	31.07	19.49
Older	156.76	233.37	37.32	16.48	18.15
<i>Load weight</i>					
5% MLC	152.16	194.82	32.23	23.86	23.07
15% MLC	164.74	231.87	38.62	23.07	16.85
25% MLC	199.64	222.13	63.52	25.13	16.85
<i>Shelf height</i>					
Wrist	261.21	161.17	50.42	20.59	13.52
Elbow	101.97	210.51	48.47	22.41	15.65
Shoulder	149.73	289.25	33.47	29.81	28.80
<i>Overall</i>	172.04	216.82	44.66	23.99	18.84

Table 8.3 Correlation coefficients between actual and predicted joint angular trajectories

	Elbow	Shoulder	Hip	Knee	Ankle
<i>Age</i>					
Younger	0.95	0.95	0.96	0.99	0.91
Older	0.96	0.95	0.98	0.98	0.81
<i>Load weight</i>					
5% MLC	0.96	0.95	0.98	0.99	0.85
15% MLC	0.96	0.95	0.97	0.98	0.85
25% MLC	0.95	0.95	0.96	0.98	0.88
<i>Shelf height</i>					
Wrist	0.92	0.94	0.96	0.98	0.90
Elbow	0.98	0.95	0.97	0.98	0.89
Shoulder	0.98	0.97	0.98	0.99	0.77
<i>Overall</i>	0.96	0.95	0.97	0.98	0.86

The weightings for younger group: $w_E = 0.5$ and $w_S = 0.5$; the weightings for older group: $w_E = 0.1$ and $w_S = 0.9$

8.5 Discussion

The major purpose of this study was to develop a novel lifting motion simulation model based on the MOO approach. A new combination of performance criteria, minimum physical effort and maximum load motion smoothness, was examined by using the weighted-sum MOO approach. The same 2-D lifting trials used for the model evaluation in Chapter 7 were simulated. The comparisons between the simulated and actual motions were made by using their absolute joint angle errors for evaluating the performance of this novel model. The optimal weightings in the MOO associated with the minimum simulation errors were determined for younger and older adults separately.

The MOO approach which integrates the two performance measures (i.e., physical effort and load motion jerk) successfully reduces the prediction errors for both younger and older age groups compared with using the single-objective optimization. In the younger group, compared to using Obj_{Effort} and Obj_{Smooth} alone in the single-objective optimization, the overall RMS absolute joint angle error decreases by 6.3% and 21.1%, respectively, when applying the optimal weightings (i.e., $w_E = 0.5$ and $w_S = 0.5$) in the MOO approach. For older adults, the overall error decreases by 28.3% and 12.1% after using the MOO approach (i.e., $w_E = 0.1$ and $w_S = 0.9$) compared with simulating motions by only using Obj_{Effort} (i.e., $w_E = 1.0$ and $w_S = 0$) and Obj_{Smooth} (i.e., $w_E = 0$ and $w_S = 1.0$), respectively. Such reduction in the overall RMS absolute joint angle error supported the argument made by Chang et al. (2001) that more than one performance criterion might be needed to better predict and explain the lifting behaviour. In addition, it also suggested that a combination of minimum physical effort and maximum load motion smoothness be an effective way to define the objective function in the MOO approach for better lifting motion prediction. Besides, in order to achieve the minimum prediction errors, the optimal weightings can be used as a reference in

future studies when using the MOO approach for lifting motion simulation with different age groups.

The age-related differences in the optimal combinations of the performance measures in the MOO approach may help understand the distinctions between younger and older adults regarding their motor control mechanisms during lifting. The physical effort is closely associated with the energy expenditure during the lifting task (Fogleman and Smith 1995); while minimizing load motion jerk during lifting is related with decreased spinal compressive force (Hsiang and McGorry 1997). For the younger adults, the optimal weightings of the physical effort and load motion jerk are both 0.5, which indicates that minimizing physical effort and reducing low back load are of equal importance when controlling the lifting motions. For older adults, given the optimal weighting values (0.1 and 0.9 for physical effort and load motion jerk, respectively), reducing low back load seems to play a more important role compared with decreasing energy expenditure in lifting motion control for older adults. Such age-related difference is consistent with our earlier findings that older adults tended to use safer lifting strategies to reduce their low back load than did younger ones (See Chapter 4-5). In fact, because of age-related degenerations in trunk muscle and intervertebral discs (Adams et al. 2002; Doherty 2003; Roughley 2004), older adults are associated with high LBP risks (Heiden et al. 2013). Therefore, older adults tend to give the first priority to reducing the low back load when planning their lifting motions. For younger adults, their lifting motion patterns (e.g., higher trunk flexion and lifting speed) can help decrease energy expenditure and increase working efficiency (reducing the total lifting duration), but induce larger low back loads. However, given their higher trunk muscle strengths (Singh et al. 2013), the increased loads might still be in the safe margin for younger adults. Therefore, younger adults choose more energy-saving and efficient kinematic patterns when planning lifting motions at the expense of increasing low back loads.

Xiang et al. (2010) also proposed a lifting motion simulation model by using the MOO approach. However, they only used the model to simulate lifting motions of young and middle-aged adults. In addition, the performance criteria used in their model appeared not good enough as they found that the addition of the maximum stability in the objective function did not significantly improve simulation accuracy. The present study is superior to Xiang et al. (2010) mainly in two aspects. First, we simulated lifting motions for both younger and older adults, and attempted to use the proposed model to identify age-related differences in lifting motion planning. As the workforce is aging worldwide and older adults tend to have higher risks of LBP, the knowledge on age-related differences in lifting motion planning is important for occupational LBP prevention. Second, the results from the experimental studies in Chapter 4 and 5, and minimizing load motion jerk is closely associated with reducing low back load (Hsiang and McGorry 1997). Thus, minimum load motion jerk was included in the objective function of the optimization procedure. These findings provide an experimental basis for the inclusion of minimum load motion jerk in the objective function of the optimization procedure which enhances the validity of our proposed model.

There are several limitations in this study. First, only 2-D sagittal lifting motions were simulated. Lifting outside the sagittal plane (i.e. asymmetric lifting) is common in industry, and associated with higher LBP risks (Davis and Marras 2005). Therefore, the present model should be further developed for simulating 3-D lifting motions in future research. Specifically, a complex 3-D human body model with more degrees of freedom will be adopted for simulating 3-D asymmetric lifting motions in our future study. Second, only two performance criteria were examined using the MOO approach in this study. Other performance criteria, such as minimum joint torque change (Uno et al. 1989) and minimum muscle force change

(Menegaldo et al. 2003), may also be investigated in the MOO approach in future studies so as to better understand the mechanisms of lifting motion planning.

In this study, a novel model for 2-D lifting motion simulation was developed by incorporating the MOO approach into the hybrid model proposed in Chapter 7. Two objective functions, minimum physical effort and minimal load motion jerk, were aggregated using the weighted-sum MOO approach. Symmetric lifting motions performed by younger and older adults were simulated. The prediction accuracy was improved for both age groups after using the MOO approach. The different optimal combinations between younger and older adults also provide implications on the age-related differences on the mechanisms of lifting motion control.

Chapter 9 Conclusions and Future Research

9.1 Research Contributions

The research presented in this dissertation investigated the age effects on lifting biomechanics from two aspects: 1) experimental studies and 2) motion simulations.

9.1.1 Experimental Investigation on Lifting Biomechanics

Existing studies (Shin et al. 2006; Burgess et al. 2009) regarding age effects on lifting biomechanics did not report low back load which is a crucial contributor to LBP incidence (Norman et al. 1998), and age-related differences were only investigated in hip and trunk kinematics. Besides, when examining age effects on postural stability, measurements of the whole-body centre of mass (COM) which provides insight into dynamic postural control mechanisms (Hof et al., 2005) were not reported. To address these limitations, age-related differences in the whole-body kinematics, low back load and postural stability were investigated under varied task conditions. Specifically, the low back load and postural stability were examined using the low back net moment and COM-COP (centre of mass – centre of pressure) inclination angles. The findings from the experiments provided implications on the development of effective interventions to reduce LBP incidences and enhance postural stability during lifting for both younger and older age groups. In particular, physical exercise programs may be a more effective ergonomic intervention for reducing the risks of LBP in lifting among older adults, compared with instructions of safe lifting strategies. As for younger workers, instructions of safe lifting strategies would be effective in LBP risk reduction. In order to maintain postural stability during lifting, especially in older adults, it is suggested the lifted load weight be adjusted according to individual strengths, and asymmetric lifting be avoided if possible.

9.1.2 Lifting Motion Simulation

Novel hybrid lifting motion simulation models using both single- and multi-objective optimization approaches were developed. These models were able to accurately predict two-dimensional (2-D) lifting motions, and well accounted for the effects of age, load weight and destination height on lifting motions. Therefore, these models could be effective ergonomics tools for evaluating and improving lifting task and workplace designs so as to prevent occupational injuries during manual handling tasks.

The proposed hybrid models are superior to the existing motion simulation models from three aspects. First, it incorporated the actual motion information in an optimization-based framework. Specifically, the joint angular velocities were constrained by the time-dependent boundaries extracted from the actual motions, so the shape of the joint angular trajectories resembled that of the actual ones. Thus, compared with the optimization-based models, this model assured the validity of the predicted motions. Second, the boundaries varied based on different task scenarios and performer's attributes (i.e., load weight, destination height and age), so the effects of these task and individual factors on the movement patterns were well accounted for in the simulation. Third, the use of the weighted-sum MOO approach substantially improves the prediction accuracy, and provides effective combinations of the objective functions for accurate lifting motion prediction for both younger and older age groups. Besides, the optimal weighting values of the performance measures derived from the MOO approach provide insights into the different mechanisms of lifting motion control between younger and older adults.

Although the model in this thesis is specifically developed for lifting motion simulation, the proposed hybrid framework for motion simulation (i.e., incorporating actual motions into the constraints of the optimization) provides the basis for constructing a universal digital human

model which is able to predict various types of human motions (e.g., reaching, walking, and etc.). In order to do so, a three-dimensional human body linkage model should be developed, and different types of real human motions are necessary for constructing a comprehensive motion database. Furthermore, this digital human model can be integrated into traditional CAD systems for proactively simulating the interaction between human beings and the virtual prototypes of various products, which enhances the product usability and reduces the cost of the whole design and manufacturing process.

9.2 Limitations and Future Research

9.2.1 Experimental Investigation on Lifting Biomechanics

There are some limitations in the experimental studies. First, foot movement was controlled in the experiment. Actually, foot movements and footstep strategies could influence trunk kinematics and risks of injury (Delisle et al. 1999; Kollmitzer et al. 2002; Sorensen et al. 2011). Second, only low back moment was examined, and spinal compression and shear forces were not used to assess low back load in the present study. Although the external low back moment has been reported as the most important contributing factor to occupational LBP (Marras 2006), spinal compression and shear forces are considered as the direct indicators for low back loading, and have been widely used to evaluate the risk of low back injuries in previous literatures (Davis and Marras 2005; Faber et al. 2008; Hoozemans et al. 2008; Kingma et al. 2004). In addition, only younger (20-30 years) and older (over 55 years) participants were recruited in the experiments. Thus, the lifting biomechanics of middle-aged subjects (31-54 years) were not reported.

Future experimental studies are needed to address these limitations. First, additional biomechanical measurements (e.g., spinal compression forces, EMG activities of trunk

muscles) should be investigated for better understanding the age-related differences in lifting biomechanics. Second, since footstep strategies may influence whole-body kinematics and low back load during lifting, age-related differences in foot movements and footstep strategies should be examined in future research. Third, to examine age-related changes in lifting biomechanics throughout the whole age range of workforce, middle-aged participants need to be recruited in the experiment as well in future studies. In addition, a randomized controlled study to evaluate the effectiveness of different interventions (e.g., lifting strategy training and physical exercise programs) in LBP prevention should be conducted for people of different age groups (i.e., young, middle-aged and older). Finally, the initial load positions were only located in the sagittal plane in the present study. Because the asymmetric initial load position would induce significant trunk lateral flexion and transverse twisting which are related with higher risks of LBP, it would be of interest to investigate age-related biomechanical differences when lifting from asymmetric load positions in future research.

9.2.2 Lifting Motion Simulation

Some limitations still exist when using the hybrid models. First, the human body is modelled using a five-segmental linkage model which is relatively simple to well account for the whole-body movements. Specifically, the pelvis, trunk and head are combined together as a rigid link, so the relative motions between these segments are neglected. Second, only 2-D lifting motions were simulated. Asymmetric lifting is common in occupations, and often associated with higher risks of LBP (Davis and Marras 2005). Finally, only two performance criteria were examined using the MOO approach in this research.

Therefore, in future research, a novel human body model which involves more degrees of freedom should be developed. For instance, the trunk can be modelled as multi-segment linkage according to the anatomical structure of the spine. Besides, the 3-D lifting motions

under varied task conditions (e.g., asymmetric load origins and destinations) should be predicted in future research. Finally, Other performance criteria, such as minimum joint torque change (Uno et al. 1989) and minimum muscle force change (Menegaldo et al. 2003), may also be investigated in the MOO approach in future studies so as to better understand the mechanisms of lifting motion planning.

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Appendix

The demographic information of each participant

No.	Gender (M/F)	Age	Body Height (cm)	Body Weight (kg)	Maximum Lifting Capacity (kg)
1	M	25	174	56	30.0
2	M	26	180	59	39.0
3	F	21	156	46	24.8
4	F	21	164	52	14.3
5	F	23	164	52	14.1
6	M	24	171	49	22.3
7	F	24	166	55	22.7
8	M	23	170	50	26.4
9	M	26	164	57	35.9
10	F	21	158	52	17.2
11	M	26	170	73	39.8
12	F	60	159	57	16.7
13	M	67	175	84	51.9
14	F	58	155	52	11.3
15	F	62	158	58	22.6
16	M	71	165	56	22.5
17	M	69	170	61	12.4
18	M	68	160	70	16.0
19	F	70	151	55	21.3
20	F	66	159	56	10.2
21	M	61	172	61	27.1
22	M	78	156	48	16.1
23	M	63	181	76	33.5

*No. 1-6 and No. 12-17 were used for model development, and No. 7-11 and No. 18-22 were used for model evaluation.

Publication List

- Song, J., Qu, X., & Chen, C.-H. (2015). Lifting motion simulation using a hybrid approach. *Ergonomics*. (In press)
- Song, J., & Qu, X. (2014). Age-related biomechanical differences during asymmetric lifting. *International Journal of Industrial Ergonomics*, 44(5), 629-635.
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- Song, J., & Qu, X. (2013). Age-related changes in trunk and knee kinematics during lifting. In *Proceedings of the Human Factors and Ergonomics Society Annual Meeting* 57(1), 934-937.
- Song, J., Qu, X., & Chen, C.-H. (2015). Age effects on postural stability in lifting tasks. In *Proceeding of the 19th Triennial Congress of the IEA* (Accepted)