THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

Pao Yue-kong Library
包玉剛圖書館

Copyright Undertaking

This thesis is protected by copyright, with all rights reserved.

By reading and using the thesis, the reader understands and agrees to the following terms:

1. The reader will abide by the rules and legal ordinances governing copyright regarding the use of the thesis.
2. The reader will use the thesis for the purpose of research or private study only and not for distribution or further reproduction or any other purpose.
3. The reader agrees to indemnify and hold the University harmless from and against any loss, damage, cost, liability or expenses arising from copyright infringement or unauthorized usage.

IMPORTANT

If you have reasons to believe that any materials in this thesis are deemed not suitable to be distributed in this form, or a copyright owner having difficulty with the material being included in our database, please contact lbsys@polyu.edu.hk providing details. The Library will look into your claim and consider taking remedial action upon receipt of the written requests.

Pao Yue-kong Library, The Hong Kong Polytechnic University, Hung Hom, Kowloon, Hong Kong

http://www.lib.polyu.edu.hk

MULTI-OBJECTIVE ANALYSIS

FOR ASSESSING THE EFFECTS OF LOAD CARRIAGE

ON THE SPINE

LI SIU WAI

Ph.D

The Hong Kong Polytechnic University

2018

The Hong Kong Polytechnic University

Department of Biomedical Engineering

Multi-objective Analysis for Assessing the Effects of Load Carriage on the Spine

LI Siu Wai

A thesis submitted in partial fulfilment of the requirements for the degree of Doctor of Philosophy

August 2017

CERTIFICATE OF ORIGINALITY

I hereby declare that this thesis is my own work and that, to the best of my knowledge and belief, it reproduces no material previously published or written, nor material that has been accepted for the award of any other degree or diploma, except where due acknowledgement has been made in the text.

LI Siu Wai

ABSTRACT

Load carriage studies have been conducted for more than a century; however, the possible risk load carriage poses on the human body is still not fully understood. Although it is difficult to determine the best way to carry a load, a backpack has its merits and comparatively more favourable human responses (in terms of subjective, physiological, kinematic, and kinetic measures) to both light- and heavy-weight carriage. However, the validity of symmetric backpack carriage employment to individuals with asymmetric body alignment, such as patients with scoliosis, remains unclear.

Most studies have evaluated various human responses to both symmetric and asymmetric load carriage merely with respect to a single objective approach and have not considered simultaneous changes in the human responses of interest. The objective of this study was to investigate the effects of continuous perturbation of an externally carried symmetric backpack carriage for healthy individuals and asymmetric single-strap cross-chest bag carriage for patients with scoliosis in static and dynamic situations. Changes in regional spinal curvature, trunk muscle activation, and lumbar spine loading were evaluated through multi-objective analysis. This was achieved by performing the present study in four interrelated parts.

The first part of the study investigated the effect of backpack carriage on the critical change in sagittal spinal curvature from the neutral upright stance in order to identify the heaviness and correctness of backpack use. This was evaluated by

assessing the spinal curvature changes along the whole spine simultaneously. Inertia-measuring sensors were used to measure the curvature changes in the cervical, upper thoracic, lower thoracic, and lumbar regions with no-load and a loaded backpack of up to 20% of body weight (BW). A multi-objective goal programming (GP) model was adopted to determine the global critical load of the maximum curvature change of the whole spine in accordance with the maximum curvature changes of the four spinal regions. The results suggested that the most critical backpack load was 13% of BW for healthy male college students.

The second part of the study evaluated the effects of carrying a backpack at 0% (no-load), 5%, 10%, 15%, and 20% of BW on the simultaneous changes in trunk muscle activation and lumbar spine loading while walking. This was investigated using an integrated system equipped with a motion analysis, a force platform, and a wireless surface electromyography (EMG) system to measure the trunk muscle EMG amplitudes and lumbar joint forces. A multi-objective GP model was developed to determine the most critical changes in trunk muscle activation and lumbar joint loading. The results suggested that lightweight backpack carriage at approximately 3% of BW might reduce the peak lumbosacral compression force by 3% during walking, compared with the no-load condition. The most critical changes in both trunk muscle activation and lumbosacral joint loading were found for a backpack loaded with 10% of BW for healthy male college students.

The third part of the study considered the effect of backpack load and boundary condition of the optimization process on the prediction of lumbar spine loading while walking towards the development of a computational algorithm in order to refine an EMG-assisted optimization (EMGAO) approach. Experimental data collected in the second part of the study were used as input data. The refined approach catered for the least possible number of variables and parameters in the optimization process and was established based on parameterized muscle gains constraining the lower boundary conditions of trunk muscle coactivations. A multi-objective GP model was developed to determine the optimal boundary condition along the backpack load spectrum between 0% and 20% of BW and a specified range of the boundary condition of the optimization process. The validity and reliability of the optimal boundary condition were analyzed using leave-one-out cross-validation and balanced bootstrap resampling methods. The refined approach provided a good estimator in terms of its unbiasedness, consistency, and efficiency for predicting the peak lumbosacral compression force.

The fourth part of the study proposed an asymmetric load carriage method for correcting spinal deformity for patients with scoliosis. Scoliosis is both a subject dependent and time-variant condition. This was investigated by employing photogrammetry to measure the simultaneous changes in scoliotic curvature in the thoracic and lumbar regions with no-load and with a properly controlled single strap cross-chest bag loaded with 2.5%, 5%, 7.5%, 10% and 12.5% of BW. Statistical tests and a multi-objective GP programming model were adopted to determine the loading conditions (placement and weight of the bag) with optimal and minimal corrections of the affected and unaffected scoliotic spinal regions, respectively. Significant short-term postural correction of scoliosis could be achieved by applying an asymmetric load on the contralateral side relative to the apex location of the major scoliotic curve. The results suggested that the application of controlled asymmetric load carriage might be a possible pragmatic method for correcting scoliotic spinal curvature. Further study of the long-term effects of subject-specific optimal asymmetric load carriage on scoliotic spinal curvatures is recommended.

In conclusion, a protocol for multi-objective analysis model was developed to investigate the effects of load carriage on the simultaneous changes in regional spinal curvature, trunk muscle activation, and lumbar spine loading during human locomotion. Such a protocol might be generalized and applied to the evaluation of other subjective, physiological, kinematic, and kinetic studies in other regions, such as the lower and upper limbs of the human body.

RESEARCH OUTPUT ARISING FROM THE THESIS

Publications

Li, S. S. W., & Chow, D. H. K. (2016). Multi-objective analysis for assessing simultaneous changes in regional spinal curvatures under backpack carriage in young adults. Ergonomics, 59(11), 1494-1504.

Li, S. S. W., & Chow, D. H. K. (2018). Effects of backpack load on critical changes of trunk muscle activation and lumbar spine loading during walking. Ergonomics, 61(4), 553-565.

Li, S. S. W., Zheng, Y. P., & Chow, D. H. K. Refined EMG-assisted optimization approach under optimal boundary condition for predicting lumbar spine loading during walking with backpack carriage. (Under revision)

Li, S. S. W., & Chow, D. H. K. (2018). Effects of asymmetric loading on lateral spinal curvature in young adults with scoliosis: A preliminary study. Prosthetics and Orthotics International. (In press)

Conference papers

Li, S. S. W., & Chow, D. H. K. (2014). A multiple criteria decision analysis model for assessing scoliotic spinal curvature under asymmetric load carriage – a preliminary study. Proceedings of the BME 2014 Biomedical Engineering International Conference, 4th – 6th December 2014, Hong Kong. (Oral presentation session: Health and Wellness Technology C-3).

Chow, D. H. K., & Li, S. S. W. (2015). A goal programming model for assessing sagittal spinal curvature under backpack carriage. Proceedings of the 25th Congress of the International Society of Biomechanics, 12th – 16th July 2015, Glasgow, UK. (Oral presentation session: Spine ISB 2015-795).

Li, S. S. W., & Chow, D. H. K. (2015). A multi-objective optimization model for assessing spinal curvature in scoliosis. Proceedings of the 25th Congress of the International Society of Biomechanics, 12th – 16th July 2015, Glasgow, UK. (Oral presentation session: Spine ISB 2015-1335).

Li, S. S. W., & Chow, D. H. K. (2017). Effects of backpack load on lumbosacral joint compression force profile and magnitude during walking. Proceedings of the 8th World Association of Chinese Biomedical Engineers World Congress on Bioengineering, 30th July – 2nd August 2017, Hong Kong. (Oral presentation session: T3.7 Biomechanics of Trunk and Upper Limb).

ACKNOWLEDGEMENTS

I sincerely thank my supervisor, Prof. Yong-Ping Zheng, for his special support, exceptional guidance, and generous tolerance throughout this project.

I wholeheartedly thank my supervisor, Prof. Daniel Hung-Kay Chow, for his dedicated encouragement, expert advice, and incredible patience throughout my research study.

I must thank Dr. Douglas Ng and Dr. Kevin Hung for their advice in numerical computation and data processing.

I wish to thank Mr. Samy Leung, Ms. Ruby Chen, and Mr. Joe Lee for their technical support as well as help in setting up and conducting the experiments.

Special thanks also give to Ms. Heidi Cheng for her support in conducting the experiments for the study of single-strap cross-chest bag carriage.

Special acknowledgement must also give to all participants recruited for this study.

Last, but not least, I would like to express my deepest gratitude to my wife Amy Ho, my daughter Edwina Li, and my son Windsor Li for their understanding and patience during my study.

TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

LIST OF ABBREVIATIONS

AIS	Adolescent idiopathic scoliosis
BW	Body weight
$C_{optimal}$	Optimal parametric gain
$C_{predictive\_optimal}$	Predictive optimal parametric gain
CFP	Compression force profile
CVGP	Chebyshev variant goal programming
DLOPT	Double linear optimization
EMG	Electromyography
EMGA	Electromyography-assisted
EMGAO	Electromyography-assisted optimization
EMG_{MVC}	Electromyography at maximum voluntary contraction
EO	External oblique
FES	Flexion and extension strategy
GRF	Ground reaction force
HNL	Head on neck lordosis
IMU	Inertia measurement unit
IO	Internal oblique
JF	Joint force
LBL	Lumbar lordosis
LBS	Load-bearing strategy
LCA	Lumbar Cobb angle

LD	Latissimus dorsi
LE	Lumbar erector spinae
LGP	Lexicographic goal programming
LSC	Local spinal curvature
LTK	Lower thoracic kyphosis
MVC	Maximum voluntary contraction
OPT	Optimization
PEA	Peak EMG amplitude
PJF	Peak joint force
PNGP	Percentage normalization goal programming
RA	Rectus abdominis
RCPEA	Rate of change of peak EMG amplitude
RCPJF	Rate of change of peak joint force
REMGAO	Refined electromyography-assisted optimization
RMSD	Root mean square difference
RRMSD	Regression of root mean square difference
SEM	Standard error of the mean
SFP	Shear force profile
SLOPT	Single linear optimization
STP	Spatial-temporal parameter
TCA	Thoracic Cobb angle
TE	Thoracic erector spinae

TLCA	Target lumbar Cobb angle
TTCA	Target thoracic Cobb angle
UTK	Upper thoracic kyphosis
WGP	Weighted goal programming
ZOGP	Zero one goal programming

LIST OF APPENDICES

CHAPTER 1

Introduction

1.1 Load carriage studies in a multi-objective approach

Load carriage is a usual daily activity of individuals. Loads can be carried in the hands or arms, or on the head, shoulders, front or back of the torso or pelvis, or around the lower limbs (Knapik et al., 2004; Legg, 1985; Motmans et al., 2006; Rose et al., 2013). The mode of load carriage depends on the convenience, weight, size, and shape of the load as well as on the duration of load carriage, culture of the terrain, physical characteristics of individuals, personal preferences, and requirements of the task-driven occupational activities or military operations (Knapik et al., 2004; Legg, 1985).

It is hard to determine a unique best way to carry a load. The general rule is to carry a load symmetrically and keep the center of mass of the load as close as possible to the center of mass of the body to maintain the load distribution of the system (body plus load) similar to the neutral upright posture in the no-load condition (Knapik et al., 2004). Backpack use is a good choice in terms of its overall and comparatively better physiological, kinematic, and kinetic measures in response to both light- and heavy-weight carriage (Legg et al., 1992; Motmans et al., 2006; Rose et al., 2013; Vieira and Ribeiro, 2015). However, excessive backpack load carriage could induce adverse effects on gait performance (Chow et al., 2005; Majumdar et al., 2013), balance and posture control (Chow et al., 2006a, 2014; Janakiraman et al., 2017), spinal repositioning ability (Chow et al., 2007, Ramsprasad et al., 2010), pulmonary capacity (Chow et al., 2009; Dominelli et al., 2012), energy expenditure (Foissac et al., 2009; Tzu-wei and Kuo, 2014), rate of muscle fatigue (Hong et al., 2008; Simpson et al., 2011a), shoulder contact pressure (Macias et al., 2005; Wettenschwiler et al., 2015), and subsequently may play a crucial role in spinal health across the full spectrum of our lifetime, especially during puberty spurt growth (Dianat et al., 2013; Hough et al., 2006).

Common guideline recommends symmetric over asymmetric load carriage. Whether this suggestion is valid for subjects with asymmetric body alignment, such as patients with scoliosis, remains unclear. When patients with scoliosis carry regular symmetric load, such as a backpack, “asymmetric” stresses are generated on their intervertebral endplates, which may induce further asymmetric spine growth and create a vicious cycle of scoliotic spinal progression (Fok et al., 2010). Because spine growth is related to applied stress (Den Boer, et al., 1999; Motmans et al., 2006), the prescription of properly controlled asymmetric loads at either side of the body is proposed for postural rectification in patients with scoliosis.

Previous studies have evaluated various human responses to load carriage with respect to a single objective approach and did not consider the simultaneous changes in physiological, kinematic, and kinetic measures. This study considered load carriage as an external perturbation and evaluated the effects of load carriage on these responses using a multi-objective analysis approach.

1.2 Objectives and hypotheses

This study assessed the simultaneous changes in regional spinal curvatures in both sagittal and coronal planes, trunk muscle activation patterns, and lumbar spine joint component forces under load carriage activities. The objectives were to investigate the effects of symmetric load carriage for healthy individuals and asymmetric load carriage for patients with scoliosis in static and dynamic situations on the changes in regional spinal curvature, trunk muscle activation, and lumbar spine loading using a multi-objective analysis approach. The hypotheses were that appropriate predictive multi-objective spine models could be identified and used to determine the critical changes in regional spinal curvature, trunk muscle activation, and lumbar spine loading during load carriage activities.

1.3 Outline of the thesis

Chapter 1 illustrates the rationale, objectives, and hypothesis of the study and briefs the contents in each chapter of the thesis.

Chapter 2 reviews relevant literature on load carriage from a historical perspective, magnitudes and modes of load carriage, backpack carriage for normal individuals, asymmetric load carriage for patients with scoliosis, and multi-objective analyses in the scope of health technology assessment.

In Chapter 3, the simultaneous curvature changes in the sagittal plane in regional spinal regions were regarded as a multi-objective analysis problem. This part of the study aimed at investigating the most critical backpack load by evaluating simultaneously the spinal curvature changes along the whole spine and hypothesized that appropriate regression and multi-objective goal programming (GP) models could be established to predict the critical regional and global spinal curvature changes between no-load and while standing with a backpack loaded up to 20% of body weight (BW).

In Chapter 4, simultaneous changes in trunk muscle activation and lumbar joint loading were considered a multi-objective analysis problem. This part of the study considered the minimum or maximum rate of change in muscle activation and joint loading as the critical event in lumbar spine strategy to allocate the minimum or maximum muscle activation and joint force per unit increase in backpack load, and evaluated the impacts of backpack weight on critical changes in trunk muscle activation and lumbar spine loading while walking. The hypothesis was that a multi-objective GP approach could be identified and applied to determine the most critical changes in trunk muscle activation (in terms of peak electromyography (EMG) amplitude) and lumbar spine loading (in terms of peak lumbosacral joint force) while walking with no-load (0% of BW) versus a backpack loaded up to 20% of BW.

In Chapter 5, the boundary conditions of biomechanical model for predicting spinal load were considered a multi-objective analysis problem. This part of the study aimed at developing a refined EMG-assisted optimization (EMGAO) approach that was minimally simple and computationally efficient in optimal boundary condition with least possible number of variables and parameters for predicting the lumbar joint loading during walking with a backpack carriage. The hypothesis was that a predictive spine model could be identified and used to determine the convergence of optimal parametric gains within a range of boundary condition for EMGAO approach along a continuous loading spectrum between no-load and a backpack loaded up to 20% of BW.

In Chapter 6, a pilot study was performed in asymmetric load carriage using a single-strap cross-chest bag for patients with scoliosis. The aim of this part of the study was to evaluate the loading configuration in both weight and position of asymmetric load carriage. The target corrective measures of both unaffected and affected scoliotic spinal curvatures in the coronal plane under asymmetric load carriage were treated as a multi-objective problem. The hypothesis was that an appropriate multi-objective GP approach could be identified and used to determine the optimal curvature changes in both the unaffected and affected regions of the scoliotic spine under properly controlled asymmetric load carriage.

In Chapter 7, synthesis of the applications of multi-objective analysis approach for assessing the simultaneous changes in spinal curvature, muscle activity pattern, lumbar spinal loading, and optimal boundary condition of an EMGAO approach for predicting spinal load was concluded and their application for future development in spine models was addressed.

CHAPTER 2

Literature review

2.1 Load carriage

2.1.1 Historical perspective

Load carriage studies have been conducted over a century, initially with target soldiers and workers as participants (Carre, 1908; Bedale, 1924). Later studies have also covered civilian users (Hale et al., 1953; Malhotra and Gupta, 1965). Researchers have been evaluating the designs of pack, modes of carriage, and limits of load, as well as psychological, physiological, and biomechanical responses to load carriage under various pacing, gradient, and terrain conditions (Winsmann and Goldman, 1976; Legg, 1985; Legg and Mahanty, 1985; Yu and Lu, 1990; Legg et al., 1992, 1997).

Other studies have investigated the heavy-load effects on the risk of injury in trampers (Lobb, 2004), movement and vigilance in soldiers (Mahoney et al., 2007), balance and decisional process in air force and army cadets (May, 2009), risk of tripping and slipping in firefighters (Park et al., 2010), psychological, physiological, and biomechanical responses in recreational hikers (Simpson et al., 2011a, 2011b, 2012a, 2012b), dynamic marksmanship in special forces soldiers (Palmer et al., 2013), static marksmanship in tactical police officers (Carbone et al., 2014), trunk and lower limb muscle activities in sports participants (Corrigan and Li, 2014), performance in mountain search and rescue personnel (Conolly, 2015), carriage efficiency in mountain porters (Bastien et al., 2016), and multi-factorial analysis of backpack-related back pain in schoolchildren (Adeyemi, 2017).

2.1.2 Magnitudes of load of carriage

In developing countries, the most cost-effective material carriage mode is head-loading. The Kikuyu and Luo women in East Africa carry load up to 70% BW on their heads with or without a forehead-supporting strap (Maloiy et al., 1986). Nepalese porters use a similar method to carry loads up to 70% of their BW (mean for women) and 146% of their BW (mean for men), and even up to 183% of their BW in rugged mountain terrain and roads (Bastien et al., 2005, 2016). Himalayan porters carry usual loads of approximately 80–90% of their BW and extreme heavy loads up to a maximum of 200% of their BW (Minetti et al., 2006) in rough terrain and steep paths up to 50% gradient in Everest mountain valley.

During military operations, ground soldiers are required to carry loads between 34 and 61 kg (40% and 71% BW based on mean BW of 86 kg) over 10–20-km distance (Knapik et al., 2004). Two recent studies in U.S. and Australian soldiers reported that the mean weights of load carriage were 45 kg (52% BW) and 48 kg (56% BW), respectively (Orr et al., 2015; Seay, 2015).

Firefighters carry essential equipment in combined loads between 30 and 40 kg (35% and 47% BW, respectively, based on mean BW of 86 kg) while performing operational duties (von Heimburg et al., 2006) that may also include a back-loading air bottle in weight of 5.4 or 9.1 kg (Park et al., 2010).

Police officers carry approximately 7 kg of vital equipment (including mobile phone, torch, first-aid kit, capsicum spray, handcuffs, baton, radio, firearm, and spare ammunition) strapped to their waist while performing general duties (Jacobsen, 2009) and up to a combined weight of 27 kg (including communication systems, body armour, and chest rigs) while performing special tactical duties (Carbone et al., 2014).

Mountain search and rescue personnel commonly carry light (11.3 kg) or standard load (22.6 kg) while performing regular or arduous duty (Conolly et al., 2015). According to the National Wildfire Coordinating Group, search and rescue personnel must be trained and pass a pack test of hiking, which covers a distance of 4.8 km, with a 20-kg pack carriage within 45 minutes (Sharkey and Gaskill, 2009).

Recreational hiking or trampering is a popular wilderness activity (Ela, 2004; Lobb, 2004). A survey conducted by Hamonko et al. (2011) in 1283 young hikers (mean BW, 69.3 kg) reported that the weight of hiking pack ranged between 9 and 39 kg (mean, 22.4 kg). Another study in trampers reported that the mean backpack load was 29% of the BW while waking for more than 5 hours over 11–15 km per day (Lobb, 2004).

Sports participants in ice hockey, baseball, soccer, football, and golf games are required to carry equipment to the field with a side-pack on one shoulder. Among those, hockey bag is comparatively heavier in weight and larger in volume. A survey among 33 young hockey players showed that the mean weight of the hockey bag was approximately 19% BW with walking distance of 642 m, and the maximum load carriage on one shoulder was up to 33% BW (Corrigan and Li., 2014). Postal workers are also required to carry a mailbag unilaterally on one shoulder. A study reported that postmen were required to a carry mailbag of 15.9 kg (Bloswick et al., 1999) and another study reported that postmen were trained by carrying a mailbag of 17.5% BW to simulate the task of performing daily postal working activities (Fowler et al., 2006).

Children in both developing and developed countries commonly carry backpacks to and from school during schooldays (Jayaratne et al., 2012; Whittfield et al., 2001). Their backpack loads are comparatively less than the load magnitudes of the aforementioned cases. Studies have reported that the weight of backpack carriage by schoolchildren ranged between 1.3 and 20.6 kg or 5.5% and 50% BW with means between 2.1 and 10.9 kg or 5.5% and 25.3% BW (Barkhordari et al., 2013; Brzek et al., 2017; Dianat et al., 2013; Dockrell et al., 2013; Forjuoh et al., 2003; Ibrahim, 2012; Lavigne, 2014; Negrini et al., 1999; Olmedo-Buenrostro et al., 2016; Shamsoddini et al., 2010). A study in college students reported that the mean weight of backpack carriage was 5.2 kg and approximately 70% of the students carried a backpack less than 10% BW (Heuscher et al., 2010). The recommended weight limits of backpack carriage range were between 5% and 20% BW (Dockrell et al., 2013).

2.1.3 Modes of load carriage

Loads can be carried in the hands or arms, or on the head, shoulders, front or back of the torso or pelvis, or around the lower limbs. The mode of load carriage depends on the convenience, weight, size, and shape of the load as well as the duration of load carriage, culture of the terrain, physical characteristics of individuals, personal preferences, and requirements of the task-driven occupational activities or military operations (Knapik et al., 2004; Legg, 1985).

For simple or convenient handling, individuals usually carry loads in their hands or arms intermittently over a short distance. Although carrying light-weight shopping bags in one hand or two hands might not have negative impacts on postural stability during standing and walking (Bampouras and Dewhurst, 2016,), this way of load carriage can cause fatigue rather quickly and is thus not recommended for heavy load, continuous, or long-distance travel (Legg, 1985).

Head-loading is commonly employed in rural areas of developing countries. However, it is not recommended because of the predominant concern of its association with the pain and discomfort induced in the neck as well as the sustainability of carrying heavier loads when compared with back-loading (Lloyd et al., 2010). Both shoulder- and back-loading could carry heavy loads, but shoulder-loading could not sustain for a longer period when compared with back-loading (Legg et al., 1992). Moreover, lower limb-loading resulted in higher energy expenditure when compared with back-loading (Abe et al., 2004).

Compared with backpack carriage, light-weight waist-pack carriage (4.5 kg) had no significant load effects on energy expenditure, perceived exertion, and lower limb postural adjustments at heel strikes (Madras et al., 1998), but heavy waist jacket carriage (>12 kg) could maintain postural sway, while increase in backpack weight significantly increased postural sway (Rugelj and Sevšek, 2011).

In accordance with the physiological, postural, kinematic, and kinetic responses, symmetric load carriage was recommended in comparison with asymmetric load carriage by one hand or a single-strap shoulder bag (Drzał-Grabiec, 2015; Ikeda et al., 2008; McGill, et al., 2013; Ozgül et al., 2012; Vieira and Ribeiro, 2015). Double pack (frontpack and backpack) was superior to front or backpack (Kinoshita, 1985; Motmans et al., 2006), and backpack was preferred to front pack (Chow et al., 2011a; Motmans et al., 2006). However, the deficiencies of double pack were the constraints of inhibiting body movement and limiting body evaporation, hindering front visual view as well as donning and doffing inconvenience (Legg, 1985; Motmans et al., 2006).

It is hard to determine a unique best way to carry a load. The general rule is to carry a load symmetrically and keep the center of mass of the load as close as possible to the center of mass of the body to maintain the load distribution of the system (body plus load) similar to the neutral upright posture in the no-load condition (Knapik et al., 2004). Backpack use is a good choice in terms of its overall and comparatively better physiological, kinematic, and kinetic measures in responses to both light- and heavy-weight carriage.

2.2 Backpack carriage

2.2.1 Health and back pain

Studies have investigated the effect of backpack load (between 0% and up to 35% BW) on health while standing and walking. Excessive backpack load (beyond certain threshold) significantly increased the amplitude of R wave of electrocardiogram (Atreya et al., 2010; Stuempfle et al., 2004), heart rate (Devorey et al., 2007; Hong et al., 2000; Stuempfle et al., 2004), blood pressure (Hong et al., 2000), metabolic cost in terms of oxygen consumption (Dames and Smith, 2015; Foissac et al., 2009; Hong et al., 2000; Lloyd and Cooke, 2000; Stuempfle et al., 2004), perceived discomfort or rate of perceived exertion (Devorey et al., 2007; Kirk and Schneider, 1992; Kistner et al., 2012; Mackie and Legg, 2008; Madras et al., 1998; Stuempfle et al., 2004), and restrictions of pulmonary function in terms of forced vital capacity and forced expiratory volume in 1 second, and peak expiratory flow (Chow et al., 2009; Legg and Cruz, 2004; Stuempfle et al., 2004).

Golriz and Walker (2011) performed a systematic review of the literature from 1966 to February 2010 on the correlation between backpack use and pain. Twenty qualified and relevant articles were selected and studied. It was found that half of them observed significant association (Grimmer and Williams, 2003; Haselgrove et al., 2008; Korovessis et al., 2004; Moore et al., 2007; Navuluri and Navuluri, 2006; Puckree et al., 2004; Sheir-Neiss et al., 2003; Siambanes et al., 2004;

Talbott et al., 2009; van Gent et al., 2003), while the other half reported that it was not statistically valid (Al-Hazzaa, 2006; Chiang et al., 2006; Goodgold et al., 2002; Heuscher et al., 2010; Iyer, 2001; Korovessis et al., 2005; Macias et al., 2005; Negrini and Carabalona, 2002; Negrini et al., 2004; Onofrio et al., 2012; Wall et al., 2003; Whittfield et al., 2005; Young et al., 2006). The findings in the relationship between backpack use and pain were still inconclusive.

A survey on school material carriage ergonomics of 1,670 Grades 6–8 schoolchildren in Sri Lanka significantly showed that there was unhealthy backpack carriage culture in developing countries that was related to health issues (Jayaratne et al., 2012). Backpack carriage was an influential factor in perceived back pain (Adeyemi et al., 2017; Golriz and Walker, 2011; Hamzat et al., 2014; Mwaka et al., 2014; Paušić et al., 2013; Vidal et al., 2013).

<u>Short note on health and back pain</u>

Excessive backpack load increases heart rate, blood pressure, energy expenditure, and perceived discomfort or rating of perceived exertion as well as restricts pulmonary capacity. Although the effects of backpack load on pain on the shoulders, neck as well as upper, mid, and lower back are inconclusive, there may be an association between the pain and load carriage during different stages of life.

2.2.2 Posture and balance

In the 1990s, Vacheron et al. (1999) conducted a primary study on the contour changes of the spine due to carriage load. The study recruited 12 subjects who were mountain guides. They were attached with markers on their spinal contour, shin, and the external occipital, and loaded with 22.5-kg pack with center of gravity at the T9 level. A TV system was used to capture the contour of the spinal curvature and measured their intervertebral disc mobility during each step. An increase in mobility in the lumbar region and a decrease in mobility in the thoracic region were significantly validated. A proper position of the spinal curvature might be adopted to sustain the 22.5-kg load carriage.

Chansirinukor et al. (2001) investigated 13 students using photography to measure their cervical and shoulder position angles under load conditions in different periods and specific percentages of BW. It was found that both time and load conditions had significant impacts on the shoulder and cervical spine posture, and increased the forward posture of the head. The recommendation was not to carry backpack weight of more than 15% of BW to maintain the proper posture of adolescents. Grimmer et al. (2002) further studied the best center location (centrally) of the backpack by investigating 250 high-school students with different backpack load on the spine at the T7, T12, or L3 level. They found that there were no significant gender, age, or load (up to 10% BW) effects, but there was a significant effect due to the factor of backpack location at the spinal profile.

Backpack should be maintained at the hip or waist level to have better control over the postural stability.

Orloff and Rapp (2004) investigated the effect of load carriage on spinal curvature. Twenty-five females were instructed to carry backpack of 9 kg and walked around a 200-m circuit at the speed of 1.79 m/s for 21 minutes. The displacement of the backpack from the spine was measured by spring-loaded rods. Displacement data were collected at 60 Hz during fatigue and rested conditions. A multivariate analysis of variance procedure was applied to analyze the data and it was found that the cubic spline curvature of the spine increased significantly. This study validated the negative effect of load carriage on the spine and the importance of the spinal curvature performance on health-related studies of the spine.

Hough et al. (2006) studied the influence of load carriage on the spine of Grade 5 and Grade 11 schoolchildren. While carrying a school bag on the back, more Grade 11 students' postures showed significant deviation from normal in the posterior area and more Grade 5 students' postures showed significant deviation from normal in the lateral area. Carriage of school bag was validated as an external effect on the developing spine. Smith et al. (2006) studied on the effects of backpack carriage during walking on the pelvic rotation, obliquity, and tilt in 30 female college students, with carriage weight under 15% of BW. A repeated measure of analysis of variance was performed and validated that there was a significant increase in pelvic tilt, but no significant effect on the pelvic obliquity or rotation. They found that carriage of backpack could have a permanent effect on the posture deviations in female college students.

Mackie and Legg (2007) recruited 10 students from the Northern, Western, Southern, and central areas of Auckland city in New Zealand and measured the time intervals and number of events of carrying school bags over a period of 24 hours. Mean carrying time and number of events of carrying school bag were found to be 119 minutes and 15 events, respectively. These data provided additional information for further studies on the effects of school bag carriage for students over simulated school-day activities. Studies of backpack carriage have focused on the effects of physical performance of the spine. Chow et al. (2007) studied the direct effects on the spinal curvature of 15 schoolchildren carrying (10, 15, and 20% of BWs) and not carrying school bags during normal upright stance. Carrying a school bag significantly reduced the upper thoracic kyphosis and lumbar lordosis and caused an immediate effect on the changes of spinal curvature and repositioning consistency.

The effect of axial load on the sagittal plane spine profile was studied by Meakin et al. (2008) to validate the variation in spine behaviour under different lifting loads. A total of 24 subjects were axial loaded with 0.8 and 16 kg across the shoulder in the upright position. A magnetic resonance imaging scanner was used to capture the images of the lower thoracic and lumbar spine. For those with smaller spinal curvature, the load made the curvature straightened, whereas for those with larger spinal curvature, the load increased the spinal curvature. Mackie and Legg (2008) examined the simulated school-day schedule of 16 boys and found that there was no significant effect on the posture of the boys if the load was limited to 10% of BW, and school bags weighing more than 10% of BW was not recommended.

Chow at al. (2010, 2011a) and Atreya et al. (2010) investigated the effects of loading, design, and carrying durations of the backpacks on the spine. Their respective findings were that spinal compression was related to carrying duration, spinal curvature was associated with spinal load conditions, and backpack design had potential effects on load carriage.

Chow et al. (2011b) evaluated the carry-over effect of backpack carriage. In their study, 13 healthy adults with backpack were allowed to walk on a treadmill for 30 minutes. They found that backpack carriage of 10% BW significantly reduced the posterior pelvic tilt and lumbar lordosis, and increased trunk forward lean, cervical lordosis, trunk kyphosis, reposition ability, and restoration of trunk posture. Even after the backpack was removed, the persistent changes in both repositioning ability and spinal curvature revealed an increased risk of spinal injury.

Rodriguez-Soto et al. (2013) investigated the effect of load carriage in lumbar spine kinematic by studying the kinematic behaviours of 10 active-duty marines. Sagittal T2 magnetic resonance images of the lumbar spine in different conditions (without load, immediate and 45 minutes after load, and walking with load) were evaluated. Inferior and superior lumbar spine levels exhibited different kinematic behaviours, and validated the lumbar flexion and postural to maintain.

Strube et al. (2017) evaluated the effect of 16.0- and 20.5-kg loads on the postural sway, trunk inclination in the sagittal plane as well as pelvis movement in the frontal and sagittal planes in 20 army cadets (mean weight: 78.8 kg), and reported that both loading conditions significantly increased mean sway velocity but did not significantly change the trunk and pelvis kinematics.

By and large, increasing backpack load increased head and trunk forward lean (Brackley et al., 2009; Chansirinukor et al., 2001; Goh et al., 1998; Devorey et al., 2007; Chow et al., 2007; Grimmer et al., 2002; Hande et al., 2012; Kistner et al., 2012; Pascoe et al., 1997) and craniovertebral, head on neck, and head and neck on trunk angles (Brackley et al., 2009; Chansirinukor et al., 2001; Chow et al., 2007; Kistner et al., 2012; Ramprasad et al., 2010) as well as decreased lumbar lordosis and thoracic kyphosis (Chow et al., 2007). Moreover, increasing backpack load increased reposition error (Chow et al., 2007, 2010); sway area; path length of center of pressure (Heller et al., 2009; Pau and Pau, 2010); displacements and velocities of both mediolateral and anterior–posterior center of pressure (Zultowski et al., 2008); and forward displacement at hip, thigh, knee, and ankle (Grimmer et al., 2002). Janakiraman et al. (2017) conducted a systematic review on the effects of backpack load on postural deviation among schoolchildren. They screened 894 articles published between February 2014 and June 2014, identified 293 related papers for full-text evaluation, including 12 high-quality papers in the systematic review, and concluded that backpack carriage of 10–15% BW is recommended for schoolchildren because carrying a backpack load exceeding 15% induced significant postural changes that might negatively affect the healthy spine. Vidal et al. (2013) investigated the effects of a 6-month postural education program on school backpack habits related to low back pain in 137 children aged 10.7 $\pm$ 0.67 years. It was found that children could learn the healthy habit of carrying a backpack, which might prevent future lower back pain.

Short note on posture and balance

Carrying a loaded backpack increases forward lean of the head, trunk, and lower limbs, as well as head on neck lordosis, postural sway, and reposition error but decreases craniovertebral angle, thoracic kyphosis, and lumbar lordosis.

2.2.3 Gait pattern

Pascoe et al (1997) studied the effect of backpack carriage on gait kinematics. Ten youths aged between 11 and 13 years were instructed to perform one stride. The dynamic conditions were captured by the TV system. There was a significant effect of load carriage on the gait pattern. Subjects promoted forward lean of trunk and head when performing the stride. Chow et al. (2005a) further investigated the load carriage effects on the movements of the pelvic, hips, knees, and ankles. They performed a repeated measures of analysis of variance procedure and validated that increase in backpack load would decrease the walking speed;

decrease pelvic motion; increase hip motion; and increase the demand of loading on hips, knees, and ankles. Further study on the effects of load carriage on gait pattern indicated that carry loads should not be exceeding 15% BW (Devroey et al., 2007).

Pau and Pau (2010) further revealed that there was a significant relation between sway area during walking and load conditions in 447 6–10-year-old schoolchildren. The sway parameters including sway area and different displacement in the anterior–posterior and medial–lateral directions were recorded. There was a significant effect of load carriage on the risk of unintentional falls and balance impairment; moreover, there was a linear relationship between backpack load and sway area. Numerous studies have investigated the effect of backpack load on spatial–temporal parameters of walking. Some studies have observed no significant changes with backpack loaded between 5% and 30% BW (Dames and Smith, 2015, Devorey et al., 2007; Goh et al., 1998; Majumdar et al., 2013), while others reported that increasing the backpack load increased double-support duration and decreased cadence, walking speed, and stride length (Birrell et al., 2009; Chow et al., 2005a; LaFiandra et al., 2003; Majumdar et al., 2010). In general, increasing backpack load increased hip and knee excursion (Birrell et al., 2009; Lafiandra et al., 2003; Liew et al., 2015), and trunk flexion (Devorey et al., 2007; Goh et al., 1998), adduction/abduction, and rotation of the hip and pelvis tilt (Birrell et al., 2009) as well as decreased range of motion of pelvic rotation and pelvic anteversion (Devorey et al., 2007). Moreover, increasing load increased peak hip adduction and range of motion of the hip in the sagittal plane as well as decreased range of motion of the pelvis in the coronal and transverse planes (Chow et al., 2005a; Lafiandra et al., 2003; Majumdar et al., 2010); however, no significant effect on the ankle joint kinematics was observed (Chow et al., 2005a).

<u>Short note on gait pattern</u>

Carrying a loaded backpack increases double-support phase and may reduce swing phase, cadence, walking, and stride length; increases trunk flexion, hip range of motion in the sagittal plane, and reduces pelvis range of motion in the coronal and transverse planes; increases peak knee flexion during the loading response phase and may have no effect on ankle kinematics.

2.2.4 Muscle activity

<u>Standing</u>

Devorey et al. (2007) investigated neck and shoulder, abdominal and back, and thigh muscle activities (in terms of mean linear envelop EMG amplitude) in 20 young adults (12 males and 8 females, mean age of 23.9 years, and body mass of 69.4 kg) with no load and with a backpack loaded at various weights (5%, 10%, and 15% BW). The effect of backpack load was significant on the activities of abdominal and back muscles but not significant on sternocleidomastoid, trapezius, rectus femoris, and biceps femoris. Increasing backpack load was found to increase rectus abdominis and external oblique, and decrease erector spinae activities. The significance was noted with backpack loads of 15%, 15%, and 10% BW, respectively.

Motmans et al. (2006) evaluated abdominal and back-muscle activities (in terms of mean EMG amplitude) in 19 college students (9 males and 10 females, mean age of 20.1 years, and body mass of 70.5 kg) between no load and backpack carriage. A backpack loaded at 15% BW significantly increased rectus abdominis and decreased erector spinae activities.

Al-Khabbaz et al. (2008) studied the effect of various loading conditions (no load and backpack load at 10%, 15%, and 29% BW) on abdominal, back, and thigh muscle activities (in terms of mean EMG amplitude) in 19 university students (mean age of 21 years and body mass of 59.7 kg). The effect of backpack load was significant on the activity of rectus abdominis and not significant on erector spinae, vastus medialis, and biceps femoris.

Gait

Harman et al. (2000) evaluated neck and shoulder, trunk, thigh, and leg muscle activities (in terms of mean root mean square EMG amplitude) in 16 male soldiers (mean age of 30.3 years and body mass of 76.8 kg) along a 15-m walkway with a backpack loaded at various weights (6, 20, 33, 47 kg) with a greater range (on average of 8%, 26%, 43%, and 61% BW). The effect of backpack load on trapezius, erector spinae, quadriceps, and gastrocnemius was significant but not for the hamstring and tibialis anterior activities. Increasing backpack load increased trapezius, quadriceps, and gastrocnemius activities. The significance occurred at 20, 33, and 33 kg, respectively. For erector spinae, increasing backpack load from 6 to 20 kg decreased the activity and from 20 to 33 kg and 47 kg increased the activity. The significance was between 20 and 47 kg backpack load. Hamstring and tibialis anterior activities increased at 20, 33, and 47 kg when compared with 6-kg backpack load.

Devorey et al. (2007) investigated neck and shoulder, trunk, and thigh muscle activities (in terms of mean linear envelop EMG amplitude) in 20 young adults (12 males and 8 females, mean age of 23.9 years, and body mass of 69.4 kg) for a 5-minute walk with no load and with a backpack loaded at various weights (5%, 10%, and 15% BW). The effect of backpack load was significant on the activities of abdominal and back muscles but not significant on sternocleidomastoid, trapezius, rectus femoris, and biceps femoris. Increasing backpack load increased rectus abdominis and external oblique, and decreased erector spinae activities. The significant association occurred at a backpack load of 10% and higher when compared with the no-load condition.

Kim et al. (2008) investigated neck and shoulder muscle activities (in terms of mean EMG amplitude) in 15 children (10 boys and 5 girls, mean age of 10.3 years, and body mass of 33.6 kg) with no load and backpack carriage during a 5-minute walk. Backpack load of 15% BW significantly increased the activities of upper trapezius, sternocleidomastoid, and midcervical paraspinals.

Hong et al. (2008) investigated neck, shoulder, and abdominal muscle activities (in terms of integrated EMG amplitude) in 15 male children (age of 6 years) for a 20-minute walk with no load and a backpack loaded at 10%, 15%, and 20% BW. Activities of lower and upper trapezius significantly increased at the thresholds of 15% and 20% BW, respectively. No significant increase in muscle activity of rectus abdominis up to 20% BW was observed.

<u>Short note on muscle activity</u>

Carrying a light- to medium-weight backpack increases abdominal muscle activities and reduces the back-muscle activities, but has no effects on neck, shoulder, and lower limb muscle activities. Carrying a heavier weight backpack further increases abdominal muscle activities and reverses the reduction to increment trend of back-muscle activities as well as increases the neck, shoulder, posterior thigh, and anterior leg muscle activities.

2.2.5 Joint loading

Chow et al. (2007) and Harman et al. (2000) studied the effect of backpack load on lower limb joint kinetics. Increasing load increased the ankle, knee, and hip joint loads and moments. Goh et al. (1998) investigated the effect of backpack load on peak lumbosacral joint forces while walking and reported that when compared with the no-load condition, 15% and 30% BW backpack increased joint compression force by 27% and 64% as well as increased resultant shear force by

25% and 61% BW, respectively. There is a lack of studies on joint loads, especially in the spinal regions compared with those studies in subjective, physiological, and kinematic responses.

<u>Short note on joint load</u>

Carrying a loaded backpack increases lower limb joint loads linearly and spinal load disproportionally. The rate of change of spinal load (per unit increase in backpack load) increases from light to medium backpack load.

2.3 Asymmetric load carriage for patients with scoliosis

2.3.1 Scoliosis

Scoliosis is a three-dimensional deformity of the spine (Weinstein et al., 2008). Whether the scoliosis is congenital, neuromuscular, or idiopathic, it progresses most rapidly during adolescent growth (Little et al., 2000). According to a survey by the Student Health Service of the Department of Health in Hong Kong, the prevalence of adolescent idiopathic scoliosis (AIS) in primary schoolchildren was 2.7%, which is the most common structural deformity in schoolchildren (Luk, 1997). The spinal deformity was demonstrated to be associated with poorer pulmonary function (Boyer at al., 1996; Muirhead and Conner, 1985) and higher prevalence of back pain (Weinstein et al., 2003), as well as motor control impairment (Nault et al., 2002; Byl et al., 1997; Sahlstrand et al., 1978) and psychosocial problem (Weinstein et al., 2003; Mayo et al., 1994).

AIS is a complex spine deformity in three-dimensional space involving lateral deviations in the frontal plane, abnormal curve profile in the sagittal plane, as well as the torsion and rotation in the transverse plane (Villemure et al., 2004). The construction of the spine in continuous curvatures in the sagittal plane is crucial for maintaining an erect, totally vertical, bipedal position (Roussouly and Pinheiro-Franco, 2011). A balance posture is obtained when the spine sacropelvis is aligned, so that horizontal gaze is maintained and the energy expenditure is minimized (Mac-Thiong et al., 2011). The sagittal global balance is vital and supported by previous studies, which validated its significant relationship with health-related quality of life in spinal deformity (Mac-Thiong et al. 2009; Glassman et al., 2005). Studies have validated that thoracic kyphosis, lumbar lordosis, as well as spinal posture in terms of pelvic tilt have significant effects on vertebral loading (Bruno et al., 2012; Bae and Mun, 2010).

The unknown aetiology and complex mechanism of the scoliosis spine growth are the challenges of the prognostic and preventive treatments of scoliosis deformities (Stokes, 2007). There is no generally validated scientific theory explaining the causes of AIS (Stokes et al., 2006). Studies have revealed that scoliosis is the result of asymmetric growth of the spine due to different growth rates of the posterior and anterior components of the spine (Pal, 1991; Sevastir et al., 1984; Roaf, 1966), and asymmetric loading on the spine that, by 'Hueter–

Volkmann Law', causes different bone growth rates within each vertebra (Stokes, 1994; Arkin and Katz, 1956), and in turn results in increase in asymmetry loading. The 'vicious cycle' continues until it is halted naturally or rectified by external means (Lafortune et al., 2007; Stokes, 2007; Stokes et al., 2006; Villemure et al., 2002, 2004). Different modulation models were developed to simulate the AIS growth (Fok et al., 2010; Stokes, 2007; Villemure et al., 2004). Validations of these growth modulation models have depicted the pathway towards the advancement study of the scoliosis, kyphosis, or even natural spine growth patterns in adolescents under different load carriages during their school life.

2.3.2 Asymmetric load carriage

Conventionally, it is believed that abnormal external loading is one of the possible factors that may exacerbate spinal deformity. Thus, children are usually recommended to carry the load symmetrically over the spine to minimize lateral trunk shifting (Negrini and Negrini, 2007; Pascoe et al., 1997) and asymmetric muscle activation observed during asymmetric carriage (Motmans et al., 2006). The shifted trunk and unilateral muscle contraction were thought to have imposed irregular biomechanical and physiological stresses on the spine, which might affect the growth of their spine. However, as asymmetric and side-shift exercises have been demonstrated to be effective physiotherapy exercise for scoliosis management (Den Boer et al., 1999; Durmala et al., 2003; Fusco et al., 2011; Mamyama et al., 2001; Maruyama et al., 2003; Mooney and Brigham, 2003; Mooney et al., 2000) and the growth of the spine was found to be associated with the applied stress (Stokes, 2002; Stokes et al., 2002), a study on the application of properly controlled asymmetric stress to the spine was proposed for postural correction and muscle conditioning in AIS. A small degree of improvement for trunk alignment might contribute to a significant biomechanical effect on a scoliotic curve, particularly for children during the rapid peripubertal growth spurt. The research hypothesis of this proposed study is that an asymmetric load carriage may serve as a physical intervention for the prevention of curve progression in scoliosis if the load magnitude and placement are properly controlled.

2.4 Multi-objective analysis

2.4.1 Decision analysis

A decision analysis that illustrates more than one requirement, criterion, or goal is referred to as a multi-criteria decision analysis (MCDA). A goal in the context of programming formulation refers to the criterion, target, or desire level to be achieved. For a specific goal, there are three types of target level to be achieved, namely, (I) at most the target level (less is better); (II) at least the target level (more is better); and (III) exactly the target level (Jones and Tamiz, 2010). A deviational variable is introduced to identify the difference between a given solution and the target level on a specific criterion. The negative and positive deviational variables represent the difference in the achieved values less than and more than the target level, respectively. The mechanism of GP is to minimize the undesired deviational variables. For Type III goal, both negative and positive deviational variables are undesirable. For Type I and Type II goals, the positive and negative deviational variables are undesirable, respectively. To achieve the optimal solution with respect to a specific criterion, the undesirable deviational variable should be minimized. The classical optimization process attempts to find an optimal solution with respect to a unique objection function. The GP model can be applied to determine an optimal solution with respect to several target objectives. The mechanism of achieving multi-objectives can be classified into two main methods, namely, lexicographic GP with pre-emptive ordering of priority levels and weighted GP without pre-emptive ordering of the objective functions but instead with different weights on the target levels (Jones and Tamiz, 2010).

Bottoms and Barlett (1975) viewed GP as an extension of classical linear programming (CLP). The foundation of mathematical formulation of CLP was established by Dantzig (1949). However, CLP allows only a single rigid criterion for determining optimal strategy. Charnes et al. (1955) introduced the conceptual theory of GP. The mathematical formulation of GP was explicitly defined by Charnes and Cooper (1961). The modification of goal setting formulation and adaptation of the occurrence of multiple conflicting goals were implemented as a GP technique to solve a wide range of problems (Aouni and Kettani, 2001).

Ignizio (1978a) generalized the GP formulations for multi-objective analysis problems and classified them as linear GP (Lee and Lerro, 1974), nonlinear GP (Ignizio, 1978b), linear integer GP (Taylor and Keown, 1978), and linear zero-one GP (Keown and Taylor, 1978). Further generalizations of GP had been done by later studies, namely, fuzzy GP (Narasimhan, 1980), GP with intervals (Romero, 1984), polynomial GP (Deckro and Hebert, 1988), GP with satisfaction functions (Martel and Aouni, 1990), convex GP (Carrizosa and Fliege, 2002), multiresponse GP (Kazemzadeh, 2008), mixed integer GP (Delice and Gungor, 2011), and Taguchi GP (Khorramshahgol, 2014; Kamran et al. 2016).

2.4.2 Health technology assessment

MCDA is a practice to systematically evaluate a set of possible courses of action, or most probably conflicting alternatives (Durbach and Stewart, 2012). The application of MCDA for health technology assessment can be broadly categorized into three aspects, namely, value measurement models, outranking models, and goal models (Thokala and Duenus, 2012). The implementations of MCDA have spread into different aspects of health technology including allocation of healthcare resources (Wilson and Gibberd, 1990; Earnshaw and Dennett, 2003; Flessa, 2003; Oddoye, 2006; Oddoye et al., 2009; Lee and Kwak, 2011), process reengineering for healthcare systems (Kwak and Lee, 2002; Epstein et al., 2007; Liberatore and Nydick, 2008), optimal arrangement in surgical operations (Arenas et al., 2002), and prioritization of healthcare projects (Morton, 2016). Tamiz et al.

(1995) identified GP as a branch of MCDA. Specific GP applications to healthcare setting are comparatively sparse. These were health resource allocation (Tingley and Liebman, 1984), time allocation in hospital pharmacy (Ghandforoush, 1993), information resource planning for a healthcare system (Lee and Kwak, 1999), and healthcare planning in a medical assessment unit (Oddoye et al., 2009).

2.5 Summary

The possible risks load carriage poses on the human body are still not thoroughly and clearly understood. Although backpack has its merits and comparatively better human responses in terms of subjective, physiological, kinematic, and kinetic measures to both light- and heavy-weight carriage, the validity of symmetric backpack carriage employment to individuals with asymmetric body alignment, such as patients with scoliosis, remains unclear. Most previous studies have evaluated various human responses to both symmetric and asymmetric load carriage only with respect to a single objective approach and have not considered simultaneous changes in the human responses under consideration. Multi-objective analysis has been applied to the evaluation of health technology assessment. However, there is a lack of such studies in physiological, kinematic, and kinetic measures in load carriage activities, especially concerning the lumbar joint loading while walking.

CHAPTER 3*

Multi-objective analysis for assessing simultaneous changes in regional spinal curvatures under backpack carriage in young male adults

* This chapter has been published by the author of this thesis.
Li, S. S. W., & Chow, D. H. K. (2016). Multi-objective analysis for assessing simultaneous changes in regional spinal curvatures under backpack carriage in young adults. Ergonomics, 59(11), 1494-1504.

Change in sagittal spinal curvature from the neutral erect stance is an important indication of the correctness and heaviness of backpack use. As current suggestions, with respect to spinal profile, of backpack load thresholds were based on significant change in individual curvature in certain spinal region only, this study evaluated the most critical backpack load by investigating simultaneous changes in spinal curvatures along the whole spine. A motion analysis system was used to measure the curvature changes in lumbar, lower thoracic, upper thoracic, and cervical regions with a backpack loaded up to 20% of body weight. A multi-objective goal programming approach was developed for determining the global critical load of maximum curvature change of the whole spine in accordance with the maximum curvature changes of the four spinal regions.

3.1 Background

Backpack carriage covers about 90% and 80% of the schoolchildren population in developed (Whittfield et al., 2001) and developing (Jayaratne et al., 2012) countries, respectively. However, 60% of schoolchildren, 95% of parents, and 73% of teachers considered it a problem (Negrini et al., 2004). Schoolchildren claimed that their backpacks caused back pain (46%), induced fatigue (66%), and were too heavy (79%) (Negrini and Carabalona, 2002). Backpack carriage is also common among adults and college students for carrying personal belongings, computers, and books to and from offices or colleges. The heaviness of backpack carriage is a concern on the spinal health of adults, adolescents, and schoolchildren and has been investigated by previous studies (Korovessis et al., 2005; Leboeuf-Yde and Kyvik, 1998; Negrini and Carabalona, 2002; Sheir-Neiss et al., 2003; Troussier et al., 1994).

Several studies have validated the negative effects of carrying a heavy backpack on gait performance (Birrell and Haslam, 2009; Chow et al. 2005a; LaFiandra et al., 2003; Smith et al., 2006), rate of muscle fatigue (Hong et al., 2008; Motmans et al., 2006; Piscione and Gamet, 2006), balance and posture (Brackley et al., 2009; Chow et al., 2007; Heller et al., 2009; Ramprasad et al., 2010; Zultowski and Aruin, 2008), energy expenditure (Foissac et al., 2009; Hong et al., 2000; Lloyd and Cooke, 2000; Stuempfle et al., 2004), pulmonary capacity (Chow et al., 2009; Legg and Cruz, 2004), and shoulder contact pressure (Macias et al., 2005). Parents are most concerned with schoolbag carriage among all the risk factors related to back pain in schoolchildren (Lucas, 2011). The association between excessive backpack load carriage and back pain is unclear. Numerous studies found that there was no significant relationship (Minghelli et al., 2014; Onofrio et al., 2012; Trevelyan and Legg, 2011; Yen et al., 2012) while others concluded that it was statistically valid (Moore et al., 2007; Sheir-Neiss et al., 2003). A recent self-reported study in college students concluded that increasing reported usual backpack weight was associated with increased yearly prevalence of low back pain (Heuscher et al., 2010). However, increase in odd ratio of annual low back pain was not significant between backpack carriage at lower weight and that exceeding 10% of body weight (BW) (Heuscher et al., 2010). The findings between college students and schoolchildren were inconsistent.

Nonetheless, backpack carriage in upright stance induces simultaneous decreases in repositioning ability of the lower thoracic and lumbar spine, changes in sagittal spinal curvatures (Chow et al., 2007, 2011b), and thus was considered as an external perturbation that may have a negative impact on the developing spine (Hough et al., 2006). Several studies also reported the vital impact of sagittal global balance and its significant effect on the development of human spine (Glassman et al., 2005; Mac-Thiong et al., 2009, 2011). The consistent changes in repositioning ability and spinal curvature exhibited a higher risk of spinal injury even after the removal of backpack (Chow et al., 2011b). Back and spine health may therefore be adversely affected by an excessively loaded backpack than was previously thought.

Spinal curvature change is a key measure of changes in posture for load carriage studies (Orloff and Rapp, 2004). Fundamental approaches for evaluating the critical backpack load with respect to spinal curvature were based on statistically significant curvature change in the spinal region under consideration and in accordance with the hypothesis of a monotonic trend beyond the threshold of critical load. The purpose of this part of the study was to apply a multi-objective approach by considering simultaneously the changes in regional curvatures to identify the global critical backpack load subject to the curvature changes along the whole spine.

Goal programming (GP) model is a multi-objective approach to determine the best solution with respect to multiple goals. In this part of the study, the multiple objectives investigated the local critical spinal curvatures in order to ascertain a global critical loading event that might position the four local spinal curvatures with least deviations from their respective critical curvatures. The rationale of multi-objective GP could be classified into two main approaches: Lexicographic GP (LGP) (Ghandforoush, 1993) and Weighted GP (WGP) (Ghufran et al., 2015; Tamiz et al., 1998). In this part of the study, both approaches were adopted.

The back health of college students has recently been a concern, but there was a lack of studies in this area (Kennedy et al., 2008). This part of the study aimed at investigating the most critical backpack load for college students by evaluating simultaneously the spinal curvature changes along the whole spine and hypothesized that appropriate regression and GP models could be established to predict the critical regional and global spinal curvature changes between no-load and while carrying a backpack loaded up to 20% of BW.

3.2 Methods

3.2.1 Participants

A convenience sample of 10 healthy young adults was recruited from a local college (Table 3.1). The sample size was a priori estimation and validated by the post-test observation of statistical power of 0.8 at level of significance set at p=0.05. Participants were screened and those with known history of any spinal deformity of low back pain were excluded. Ethical approval from the Human Research Ethics Committee and written informed consent from all participants were obtained beforehand.

Table 3.1 Demographics of the ten healthy male participants.

Participant	Height (m)	Weight (kg)	Age (year)
1	1.76	69.3	29
2	1.77	72.6	22
3	1.67	58.1	23
4	1.81	66.7	23
5	1.80	70.8	20
6	1.71	61.2	20
7	1.67	65.3	23
8	1.71	56.5	22
9	1.71	70.1	20
10	1.78	70.2	22
Mean	**1.74**	**66.1**	**22.4**
SD	**0.05**	**5.7**	**2.5**

3.2.2 Experimental design

A two-factor randomized repeated measure design was adopted in this part of the study. The dependent variables were the four local spinal curvatures of lumbar lordosis (LBL), lower thoracic kyphosis (LTK), upper thoracic kyphosis (UTK), and head on neck lordosis (HNL). The independent variables were the loading condition (0% BW for no-load condition and 5%, 10%, 15%, and 20% BW for a loaded backpack carriage) and trial order (1^{st}, 2^{nd}, and 3^{rd}). The backpack (30 cm × 14 cm × 38 cm) used in this part of the study was an internally reconstructed standard dual strap schoolbag and made of light weight synthetic material (Figure 3.1). Its back was cut and supported by an internal frame with two metal bars and a detachable load-bearing container (Chow et al., 2007; Vacheron et al. 1999). The sensors attached to the spine could move freely without any constraints and the level of center of gravity of the backpack could be adjusted by the shoulder straps and a detachable horizontal belt.

(a) (b)

Figure 3.1 An internally reconstructed standard dual strap schoolbag. (a) The middle of its back was cut. (b) An internal frame was embedded and the level of center of gravity of the backpack was identified by a horizontal belt.

The effects of foot posture on spine biomechanics have been reported in a previous study (Hu et al., 2014). The placement of the feet during quiet upright stance was standardized for balance testing (Mcllroy and Maki, 1997) and adopted in this part of the study to reduce the possible within-subject spinal profile changes due to uncontrolled postural sway during repeated measures in various loading conditions. The participants were instructed to maintain a standardized and relaxed barefoot erect stance with an angle of 30° between the long axes of the feet, heels 10 cm apart and gaze fixed on a target (reference square of 14 cm × 15 cm) placed 2 m in front of and adjusted at their eye level, and arms hanging freely at both sides (Chow et al., 2007). Then, they were instructed to symmetrically carry the backpack loaded at 5, 10, 15 and 20% BW (Hong and Cheung, 2003). The center of gravity of the backpack was adjusted at the T12 spinal level by shortening or lengthening the shoulder straps (Chow et al., 2007; Grimmer et al., 2002; Vacheron et al., 1999). The sequence of performing the experimental trials was randomized. Three trials for each no-load and loaded conditions were acquired for 3 s in the midway during a 45-s quiet erect stance simulating the common waiting time for a transportation vehicle or elevator (Zultowski and Aruin, 2008). Participants rested for 3 min between consecutive trials to reduce any confounding effects of fatigue.

3.2.3 Data collection

This part of the study used the inertial measurement unit (IMU) developed by Noraxon (MyoMOTION, Noraxon, USA) to overcome the shortcoming of marker occlusion problem of motion capturing by traditional video-based motion analysis system. A triaxial gyroscope and a North magnetic sensor were installed in each IMU. The motion-related signals of 3-dimensional anatomical angles between the IMUs attached to identified body landmarks were transmitted to and received by data acquisition software (MR3, Noraxon, USA). The sampling frequency was 100 Hz. IMU signals were processed by a built-in Kalman filter and integrated to obtain their orientation angles. When comparing with gold standard (Vicon TV system), the accuracy of IMU in measuring body segment orientation angles was 1.27±1.3° (Balasubramanian, 2013). Five IMUs were affixed to participant's skin surface proximal to S1, T12, T7, and C7 spinous processes as well as external occipital (EO) (Figure 3.2). The 4 local spinal curvatures were recorded for 3 s for each experimental trial. The curvatures projected on the sagittal plane were considered the principle motion of the spine while carrying a backpack symmetrically and assumed the curvature changes in the transverse and coronal planes were minimal.

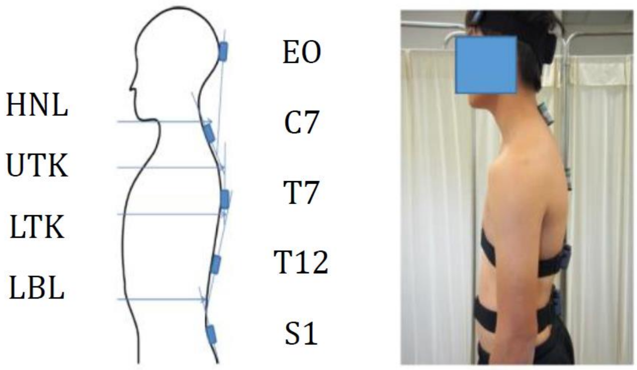


Figure 3.2 Measurements of local spinal curvatures. Five inertial measurement units (IMUs) were affixed to the spine for observing the head on neck lordosis (HNL), upper thoracic kyphosis (UTK), lower thoracic kyphosis (LTK), and lumbar lordosis (LBL) between adjacent IMUs.

3.2.4 Data analysis

A two-way repeated measure ANOVA design was adopted in this part of the study with backpack load (0%, 5%, 10%, 15%, and 20%) and trial order (1^{st}, 2^{nd}, and 3^{rd}) as within-subject factors to analyze the LBL, LTK, UTK, and HNL (SPSS version 20.0, IBM Inc., Chicago, IL, USA). Thresholds of statistical power (1-β) and significance (α) were set at 0.8 and 0.05, respectively. Post hoc pairwise comparisons were performed based on least significant difference (LSD) criterion.

Scatter plots of overall means of four local spinal curvatures against the loading condition were obtained and fitted with various regression lines. An appropriate continuous regression model for the LBL, UTK, LTK, and HNL was identified. The four local spinal curvatures (LSC) were formulated as continuous functions of backpack load (L, % of BW) as below:

$$LSC_i = f_i(L), \text{ where } i = 1(\text{HNL}), 2(\text{UTK}), 3(\text{LTK}), 4(\text{LBL}).$$

A goal in a programming approach refers to the desired, target, or criterion level to be achieved. For an identified goal, there are three types of target levels: (Type I, less is better, "≤ target level"), (Type II, more is better, "≥ target level"), and (Type III, exactly same, "= target level")(Jones and Tamiz, 2010). A deviational variable is proposed to measure the difference between the target level and a given solution on a specific requirement. Positive or negative deviational variables depict the difference in the achieved values above or below the target level, respectively. The rationale of goal programming is to minimize the undesired deviational variables in order to obtain the optimal solution with respect to a specific requirement. For Type II and Type I goals, the negative and positive deviational variables are respectively undesirable. For Type III goal, both positive and negative deviational variables are unfavorable.

Lexicographic goal programming (LGP) examined the priority levels in pre-emptive ordering. As reported in previous studies that the prevalence of back pain bore different significance in different spinal regions (neck: 30%, mid-back: 13%, and low back: 43%) (Leboeuf-Yde et al., 2012), the priority of target spinal levels was ordered as (1) lumbar lordosis, (2) head on neck lordosis and (3) lower and upper thoracic kyphosis. The solutions of the global critical backpack loads were figured out by the sequential optimization approach as stated below:

<u>LGP</u>

Minimize $[D_4^- \text{ or } D_4^+]^1,\ [D_1^- \text{ or } D_1^+]^2,\ [(D_2^- \text{ or } D_2^+) + (D_3^- \text{ or } D_3^+)]^3$

subject to $f_i(L) + D_i^- - D_i^+ = TC_i$

$5 \leq L \leq 20$

$D_4^- \text{ or } D_4^+ = OS_1$

(for the 2nd and 3rd GPs only, OS_1 was the optimal solution of the 1st GP)

$D_1^- \text{ or } D_1^+ = OS_2$

(for the 3rd GP only, OS_2 was the optimal solution of the 2nd GP)

$D_i^-,\ D_i^+ \geq 0$

Where

L represented the weight of backpack in % BW

D_i^- represented the negative deviational variable for spinal region i

D_i^+ represented the positive deviational variable for spinal region i

MC_i represented the maximum curvature level for spinal region i,

TC_i represented the target curvature level for spinal region i

Weighted goal programming (WGP), on the other hand, considered the different weights on the target levels. WGP weighted the undesirable deviations from target spinal curvatures by three different methods: Chebyshev variant (CVGP), percentage normalization (PNGP), and zero-one normalization (ZOGP). CVGP and PNGP adopted target curvatures as weights whereas ZOGP used maximum curvatures as weights. CVGP required an optimal solution with minimum value across all weighted undesirable deviations while PNGP only demanded the minimum of the total weighted undesirable deviations. The solutions of the global critical backpack loads were processed by the optimization approaches stated below:

CVGP

Minimize $\quad k$

subject to $\quad f_i(L) + D_i^- - D_i^+ = TC_i$

$$\frac{(D_i^- \text{ or } D_i^+)}{TC_i} \leq k$$

$$5 \leq L \leq 20$$

$$D_i^-, D_i^+ \geq 0$$

PNGP

Minimize $\quad \sum_{i=1}^{4} \frac{(D_i^- \text{ or } D_i^+)}{TC_i}$

subject to $\quad f_i(L) + D_i^- - D_i^+ = TC_i$

$$5 \leq L \leq 20$$

$$D_i^-, D_i^+ \geq 0$$

ZOPG

Minimize $\quad \sum_{i=1}^{4} \frac{(D_i^- \text{ or } D_i^+)}{MC_i}$

subject to $\quad f_i(L) + D_i^- - D_i^+ = TC_i$

$$5 \leq L \leq 20$$

$$D_i^-, D_i^+ \geq 0$$

The LGP, CVGP, PNGP, and ZOGP optimization problems were solved by implementing LINGO 10.0 software (Lindo Systems Inc., Chicago, IL, USA).

3.3 Results

3.3.1 Local spinal curvatures

The local spinal curvatures of the ten subjects at each backpack load and trial order were recorded (Appendix 3.1) and the results of statistical analyses of the load, trial and their interaction effects on the spinal curvatures were determined (Table 3.2). There were no significant trial as well as load and trial interaction effects on HNL (p=0.533 and 0.557, respectively), UTK (p=0.883 and 0.666, respectively), LTK (p=0.132 and 0.106, respectively), and LBL (p=0.329 and 0.703, respectively). Moreover, HNL did not significantly increase among no-load (0% BW) and backpack loaded conditions (p=0.296). However, significant changes were observed in the UTK (p=0.001, power=0.981), LTK (p=0.018, power=0.801), and LBL (p=0.017, power=0.808). Post hoc pairwise comparisons showed significant decreases in LBL and UTK under all backpack loaded conditions, and a significant increase in LTK but at 15% BW only, as compared with the no load condition. In addition, among the loaded conditions of UTK was further significantly decreased with backpack loaded from 5% to 15% BW.

Table 3.2 Statistical analyses of the spinal curvatures: head on neck lordosis (HNL), upper thoracic kyphosis (UTK), lower thoracic kyphosis (LTK), and lumbar lordosis (LBL) at each loading condition from the pooled data of all participants

<table>
<tr><th rowspan="3">Pooled Mean Spinal Curvature (degree)</th><th colspan="3">p-values of main and interaction effects (2-way repeated measure ANOVA)</th><th colspan="5" rowspan="2">Loading condition (% BW)</th></tr>
<tr><th colspan="3">Power of the test with significant effect</th></tr>
<tr><th>Trial</th><th>Load and Trial</th><th>Load</th><th>0</th><th>5</th><th>10</th><th>15</th><th>20</th></tr>
<tr><td>HNL</td><td>0.533</td><td>0.557</td><td>0.296</td><td>32.1</td><td>33.6</td><td>33.8</td><td>33.7</td><td>32.7</td></tr>
<tr><td rowspan="2">UTK</td><td>0.883</td><td>0.666</td><td>0.001*</td><td rowspan="2">32.9</td><td rowspan="2">29.1^</td><td rowspan="2">27.6^</td><td rowspan="2">27.3^#</td><td rowspan="2">27.9^</td></tr>
<tr><td colspan="3">power = 0.981</td></tr>
<tr><td rowspan="2">LTK</td><td>0.132</td><td>0.106</td><td>0.018*</td><td rowspan="2">10.3</td><td rowspan="2">12.3</td><td rowspan="2">12.4</td><td rowspan="2">12.3^</td><td rowspan="2">11.9</td></tr>
<tr><td colspan="3">power = 0.801</td></tr>
<tr><td rowspan="2">LBL</td><td>0.329</td><td>0.703</td><td>0.017*</td><td rowspan="2">17.3</td><td rowspan="2">15.4^</td><td rowspan="2">15.2^</td><td rowspan="2">15.1^</td><td rowspan="2">15.2^</td></tr>
<tr><td colspan="3">power = 0.808</td></tr>
<tr><td colspan="9">* p<0.05 (significant load effect on UTK, LTK, and LBL)
Pair-wise comparison of load effect (p<0.05):
^ significant different when comparing with 0% BW
# significant different when comparing with 5% BW</td></tr>
</table>

3.3.2 Regressions of local spinal curvatures

An appropriate continuous regression line formulating the relationship between loading condition and individual local spinal curvature was identified to be the second-order polynomial model (Figure 3.3). The R^2 of each regression model of the data set of LBL, LTK, UTK and HNL were 0.914, 0.906, 0.992 and 0.968, respectively. The four regressions illustrated relatively strong unimodal quadratic associations between loading condition and local spinal curvatures. While loading increased, both LTK and HNL rose from the baseline condition until they reached their local critical loads at 12.3% and 10.9% BW, respectively, and then reversed towards reduction with further increase in backpack load. On the contrary, both LBL and UTK reduced in the beginning of the loading until they reached their local critical loads at 14.0 and 14.1% BW, respectively, before reversing to increase backpack load.

3.3.3 Goal programming models

The maximum and minimum curvatures of LBL, LTK, UTK, and HNL were investigated and the local critical curvatures were figured out with respect to four local critical loads. The LGP and 3 WGP approaches were formulated in accordance with the respective cost functions, constraints, boundaries, and nonnegativity conditions (Tables 3.3 and 3.4). The global critical loads of the LGP, CVGP, PNGP and ZOGP were found to be 14.0, 13.1, 13.0 and 12.9 % BW respectively.

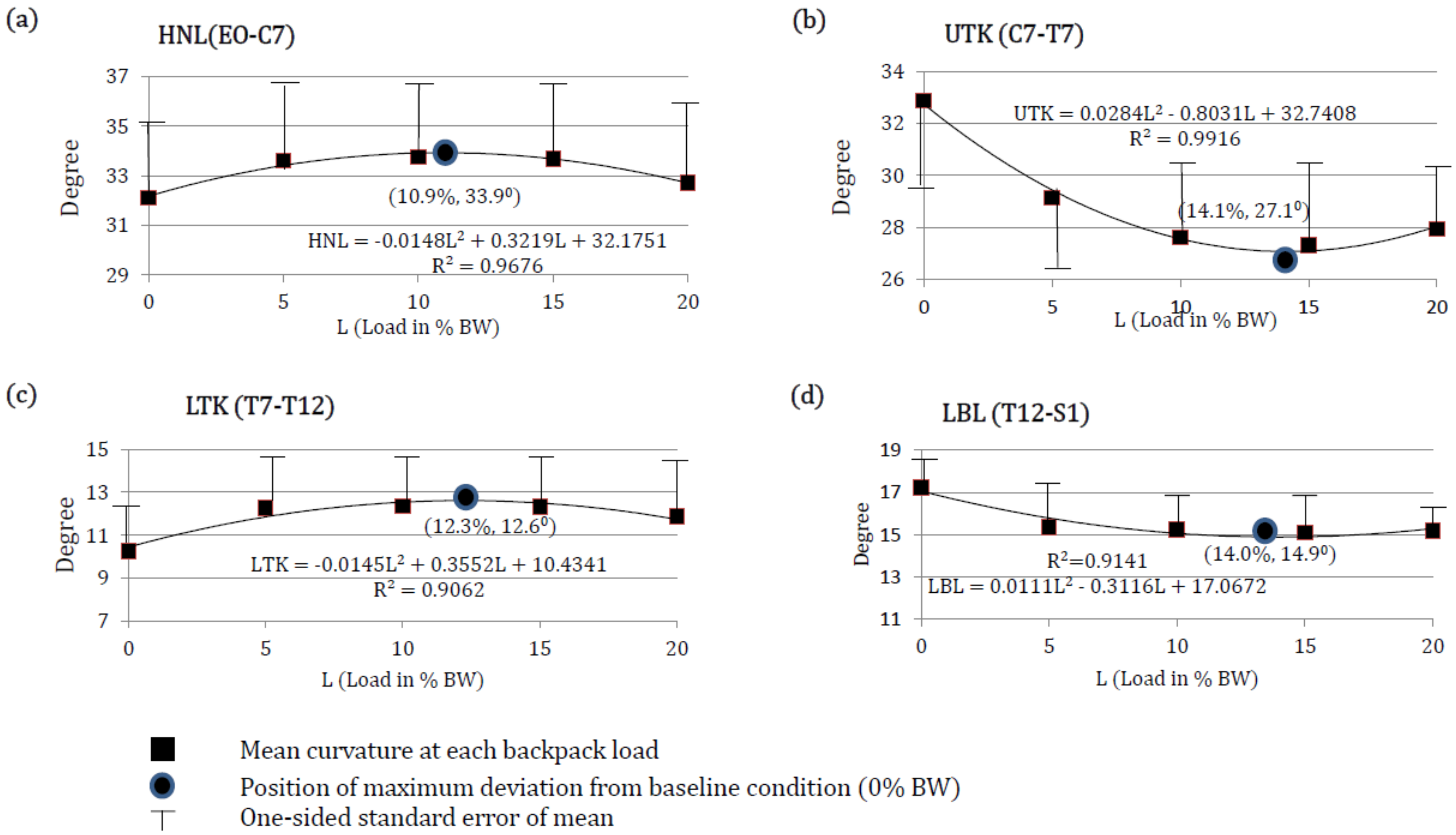


Figure 3.3 The regressions of mean angles from all pooled data: (a) head on neck lordosis (HNL), (b) upper thoracic kyphosis (UTK), (c) lower thoracic kyphosis (LTK), and (d) lumbar lordosis (LBL). The proportion of total variability explained by each regression model was measured by the R^2 shown on the figure.

Table 3.3 Formulation of the Lexicographic Goal Programming approach.

Lexicographic Goals: (1) LBL , (2) HNL, (3) LTK + UTK	
Cost Function (1st GP)	D_{LBL}^{+}
Constraint	$-0.0148L^2 + 0.3219L + D_{HNL}^{-} - D_{HNL}^{+} = 1.7503$ $0.0284L^2 - 0.8031L + D_{UTK}^{-} - D_{UTK}^{+} = -5.6775$ $-0.0145L^2 + 0.3552L + D_{LTK}^{-} - D_{LTK}^{+} = 2.1753$ $0.0111L^2 - 0.3116L + D_{LBL}^{-} - D_{LBL}^{+} = -2.1868$
Boundary	$5 \le L \le 20$
Non-negativity	$D_{HNL}^{-}, D_{HNL}^{+}, D_{UTK}^{-}, D_{UTK}^{+}, D_{LTK}^{-}, D_{LTK}^{+}, D_{LBL}^{-}, D_{LBL}^{+} \ge 0$
Cost Function (2nd GP)	D_{HNL}^{-}
Constraint	$-0.0148L^2 + 0.3219L + D_{HNL}^{-} - D_{HNL}^{+} = 1.7503$ $0.0284L^2 - 0.8031L + D_{UTK}^{-} - D_{UTK}^{+} = -5.6775$ $-0.0145L^2 + 0.3552L + D_{LTK}^{-} - D_{LTK}^{+} = 2.1753$ $0.0111L^2 - 0.3116L + D_{LBL}^{-} - D_{LBL}^{+} = -2.1868$ $D_{LBL}^{+} = 0$
Boundary	$5 \le L \le 20$
Non-negativity	$D_{HNL}^{-}, D_{HNL}^{+}, D_{UTK}^{-}, D_{UTK}^{+}, D_{LTK}^{-}, D_{LTK}^{+}, D_{LBL}^{-}, D_{LBL}^{+} \ge 0$
Cost Function (3rd GP)	$D_{LTK}^{-} + D_{UTK}^{+}$
Constraint	$-0.0148L^2 + 0.3219L + D_{HNL}^{-} - D_{HNL}^{+} = 1.7503$ $0.0284L^2 - 0.8031L + D_{UTK}^{-} - D_{UTK}^{+} = -5.6775$ $-0.0145L^2 + 0.3552L + D_{LTK}^{-} - D_{LTK}^{+} = 2.1753$ $0.0111L^2 - 0.3116L + D_{LBL}^{-} - D_{LBL}^{+} = -2.1868$ $D_{LBL}^{+} = 0$, $D_{HNL}^{-} = 0.1444576$
Boundary	$5 \le L \le 20$
Non-negativity	$D_{HNL}^{-}, D_{HNL}^{+}, D_{UTK}^{-}, D_{UTK}^{+}, D_{LTK}^{-}, D_{LTK}^{+}, D_{LBL}^{-}, D_{LBL}^{+} \ge 0$

Solution of LGP was 14.0% BW.

Table 3.4 Formulation of three Weighted Goal Programming approaches.

Chebyshev Variant Goal Programming (CVGP)	
Cost Function	h
Constraint	$-0.0148L^2 + 0.3219L + D^-_{HNL}$ or $D^+_{HNL} = 1.7503$ $0.0284L^2 - 0.8031L + D^-_{UTK}$ or $D^+_{UTK} = -5.6775$ $-0.0145L^2 + 0.3552L + D^-_{HNL}$ or $D^+_{HNL} = 2.1753$ $0.0111L^2 - 0.3116L + D^-_{UTK}$ or $D^+_{UTK} = -2.1868$ $\frac{D^-_{HNL}}{33.9254} \leq h$, $\frac{D^+_{UTK}}{27.0633} \leq h$, $\frac{D^-_{LTK}}{12.6094} \leq h$, $\frac{D^+_{LBL}}{14.8804} \leq h$
Boundary	$5 \leq L \leq 20$
Non-negativity	$D^-_{HNL}, D^+_{HNL}, D^-_{UTK}, D^+_{UTK}, D^-_{LTK}, D^+_{LTK}, D^-_{LBL}, D^+_{LBL} \geq 0$
Percentage Normalization Goal Programming (PNGP)	
Cost Function	$\frac{D^-_{HNL}}{33.9254} + \frac{D^+_{UTK}}{27.0633} + \frac{D^-_{LTK}}{12.6094} + \frac{D^+_{LBL}}{14.8804}$
Constraint	$-0.0148L^2 + 0.3219L + D^-_{HNL}$ or $D^+_{HNL} = 1.7503$ $0.0284L^2 - 0.8031L + D^-_{UTK}$ or $D^+_{UTK} = -5.6775$ $-0.0145L^2 + 0.3552L + D^-_{HNL}$ or $D^+_{HNL} = 2.1753$ $0.0111L^2 - 0.3116L + D^-_{UTK}$ or $D^+_{UTK} = -2.1868$
Boundary	$5 \leq L \leq 20$
Non-negativity	$D^-_{HNL}, D^+_{HNL}, D^-_{UTK}, D^+_{UTK}, D^-_{LTK}, D^+_{LTK}, D^-_{LBL}, D^+_{LBL} \geq 0$
Zero-one Normalization Goal Programming (ZNGP)	
Cost Function	$\frac{D^-_{HNL}}{33.9254} + \frac{D^+_{UTK}}{29.4353} + \frac{D^-_{LTK}}{12.6094} + \frac{D^+_{LBL}}{15.7867}$
Constraint	$-0.0148L^2 + 0.3219L + D^-_{HNL}$ or $D^+_{HNL} = 1.7503$ $0.0284L^2 - 0.8031L + D^-_{UTK}$ or $D^+_{UTK} = -5.6775$ $-0.0145L^2 + 0.3552L + D^-_{HNL}$ or $D^+_{HNL} = 2.1753$ $0.0111L^2 - 0.3116L + D^-_{UTK}$ or $D^+_{UTK} = -2.1868$
Boundary	$5 \leq L \leq 20$
Non-negativity	$D^-_{HNL}, D^+_{HNL}, D^-_{UTK}, D^+_{UTK}, D^-_{LTK}, D^+_{LTK}, D^-_{LBL}, D^+_{LBL} \geq 0$

Solutions of CVPG, PNGP, and ZOGP were 13.1, 13.0, and 12.9 % BW, respectively.

3.4 Discussion

3.4.1 Spinal curvature changes

When comparing between the loaded conditions (5, 10, 15, and 20% BW) with the baseline condition (0% BW, no-load), there was no significant increase in HNL but there were significant decreases in LBL and UTK at all loads and a significant increase in LTK at load of 15% BW only. These results were consistent with those reported by Chow et al. (2007) that there were significant increases in head on neck extension, significant decrease in LBL at a load of 20% BW only, significant decrease in UTK, and no significant decrease in LTK at all loads.

The changes of LBL and HNL were largely consistent among different studies, i.e. decrease in LBL and increase in HNL while carrying a loaded backpack (Chansirinukor et al., 2001; Chow et al., 2006a, 2007, 2010, 2011a; Devroey et al., 2007; Goodgold et al., 2002; Grimmer et al., 2002; Hong and Cheung, 2003; Li et al., 2003; Pascoe et al., 1997; Rodríguez-Soto et al., 2013). The decrease in LBL was the consequence of the lumbar flexion maneuver, which aimed at centralizing a posterior load over the base of support while the increase in HNL was to maintain the eye gaze with the increased trunk forward lean.

The change of thoracic kyphosis was inconsistent across various findings reported from previous studies. A falling trend was found in the studies by Chow et al. (2007) and Negrini and Negrini (2007) as well as in this part of the study (43.2°, 41.4°, 40.0°, 39.6°, and 39.8° at loads of 0, 5, 10, 15, and 20% BW, respectively by adding the UTK and LTK at each load) whilst a rising trend was observed in the other studies by Chow et al. (2010, 2011b), Devroey et al. (2007), and Orloff and Rapp (2004). The difference in findings might be accounted by the design of the carried load. A regular backpack was used in the latter studies while a specially reconstructed one (which might have constrained the movement of the thoracic spine and resulted in a relatively straightened thoracic curve) was used in the former.

3.4.2 Flexion and extension strategies

The human spine is a complex structure. It is difficult to fully understand its function (Reeves et al., 2011). Nonetheless, its mechanism may be better explained by its capability to maintain both dynamic and static equilibrium under external perturbation (Reeves et al., 2007). The flexion and extension strategies (FES) of the four spinal regions simultaneously changed with respect to the continuous perturbation by carrying a loaded backpack and revealed the mechanism of the functional spine. When backpack load increased, the local spinal curvatures could be expressed as continuums of an external perturbation.

The four spinal regions exhibited different flexion and extension strategy between backpack load of 5 and 20% BW. The flexion and extension strategy could be partitioned by the four local critical load conditions for detail investigation (Table 3.5). The respective flexion and extension of LBL, LTK, UTK, and HNL when loading increased from 5% to 10.9% BW (Stage 1) were [E−E|F−F]; from 10.9 to 12.3% BW (Stage 2) were [F|E|F−F]; from 12.3 to 14.0% BW (Stage 3) were [F|E−E|F]; from 14.0 to 14.1% BW (Stage 4) were [F|E−E−E], and from 14.1 to

20% BW (Stage 5) were [F—F|E—E]. Beginning in Stage 1 and ending at Stage 5, flexion and extension strategy of all four regions were synchronized in both upper and lower spinal regions. The difference between Stage 1 and Stage 5 was that the lower [LTK and LBL] and upper [HNL and UTK] spinal regions reversed their strategy from flexion to extension and from extension to flexion, respectively. In Stage 3, flexion and extension strategy of both lower and upper spinal regions was counter-active. Stages 2 and 4 transitioned the flexion and extension strategy from Stage 1 to Stage 3 and from Stage 3 to Stage 5, respectively.

Table 3.5 Flexion and extension strategy of individual spinal region

Spinal curvature	Flexion and Extension Strategy (FES)				
	(Stage 1) 5-10.9% BW	(Stage 2) 10.9-12.3% BW	(Stage 3) 12.3-14.0% BW	(Stage 4) 14.0-14.1% BW	(Stage 5) 14.1-20% BW
HNL (EO-C7)	E	F	F	F	F
UTK (C7-T7)	E	E	E	E	F
LTK (T7-T12)	F	F	E	E	E
LBL (T12-S1)	F	F	F	E	E

F = Flexion, E = Extension,
BW = body weight, EO = external occipital,
HNL = head on neck lordosis, UTK = upper thoracic kyphosis,
LTK = lower thoracic kyphosis, LBL = lumbar lordosis

3.4.3 Critical backpack load

The size and extent of thoracic spine articulated with the rib cage are distinctive from the cervical and lumbar spine. This unique architecture leads to a different regional movement maneuver (Willems et al, 1996). The lower and upper thoracic spine synthesizes the movement pattern, to a certain extent, of the lumbar and cervical, respectively (Cook, 2007). Aiming at maintaining stability, the spine is necessarily, at the perturbed condition under backpack load carriage, to stay close to its neutral postural position (unperturbed condition, without backpack carriage), and most importantly, constrain the ranges of motion of the four spinal regions (Reeves et al., 2007). The range of motions and their unimodal quadratic local spinal curvatures observed in this part of the study largely complied with the requirement of a healthy spine to maintain its stability under the external perturbation while carrying a backpack loaded between 5 and 20% BW.

Only symmetric load carriage between 5% and 20% BW using an internally reconstructed backpack with two-strap and center of gravity positioned at T12 level was evaluated in this part of the study. There were various factors that might have impact on the curvature changes while carrying a loaded backpack such as: the position of the center of gravity of the loaded backpack (Devroey et al., 2007), the way of carrying backpack (Chow et al., 2011a; Devroey et al., 2007), load distribution (Knapik et al., 1996), gender difference (Attwells et al., 2006), skill and experience of backpack users (Knapik et al., 2004; Vacheron et al., 1999), and the functional designs of the backpack (Holewijn, 1990).

What heaviness is excessive? The general recommendation is not to exceed certain limit of allowable backpack weight. Previous studies about limit of backpack weight had been done intensively for schoolchildren worldwide. The common guidelines, generally in multiples of 5% BW, are inconsistent across countries (Lindstrom-Hazel 2009). The range is between 5 and 20% BW (Dockrell et al., 2013). The guidelines with respect to specific postural changes or spinal curvature were consistent with the recommended guideline of 5% (Rateau, 2004), 10% (Devroey et al., 2007), 15% (Chansirinukor et al., 2001), and 20% BW (Al-Khabbaz et al., 2008).

In this study, the highest and lowest thresholds of critical backpack loads were shown by the significant changes in UTK and LBL at 5 and 20% BW levels, respectively. The most critical range of backpack load fell in the range between 13% BW (WGPs) and 14% BW (LGP), as explained by the counter-active flexion and extension strategy catered by both lower and upper spinal regions. This part of the study therefore recommends 13% BW (lower limit upon comparing the optimal solutions between LGP and WGPs) as the most critical backpack load as it falls within the most critical load region and the mostly recommended backpack weight limit of 10%–15% BW by previous studies (Dockrell et a., 2013).

3.4.4 Advantage of considering global critical load

The most critical backpack load was evaluated with respect to the global critical load of the whole spine, instead of the local critical load of a specific spinal region or threshold level. A more comprehensive and reasonable standard considering both the postural changes and spinal stability as well as their associations with the increase in backpack load carriage could be determined.

3.4.5 Limitations

There were several limitations in this part of the study. First, for convenience of conducting the experiment by a male experimenter, female participants were not included, application of the findings to females is subject to further validation. Second, as all participants were young adults, the findings may not be generalizable to adolescents and children. Third, the small standard deviations in height (0.05 m) and body weight (5.7 kg) might exhibit homogeneousness of the ten randomly selected participants and constrain the generalization of the findings of this part of the study as well as integrate possible over-fitting of experimental data to the recommended second-order polynomial regression model. Fourth, the employment of an internally reconstructed backpack with center of gravity positioned at the T12 level might be different from regular backpack use. Fifth, this part of the study aimed at evaluating the short-term effect of backpack load carriage and the findings might not be applicable to long-term carriage. Sixth, muscle fatigue and physiological responses including ECG, heart rate, blood pressure, oxygen consumption, and pulmonary functions was not considered in short-term carriage. Seventh, the current findings did not consider the parameters of spinal flexibility and muscle strength of individual participants. In addition, this part of the study compared the curvature changes partitioned in four different spinal regions and determined the most critical backpack load by LGP and WGPs.

3.5 Summary

This part of the study applied inertia-measuring sensors to assess simultaneous changes in local spinal curvatures along the whole spine in upright stance with backpack carriage up to 20% of BW. A multi-objective goal programming approach was developed to determine the global critical loading condition. Results suggested that for healthy male young adults, the most critical backpack load was 13% of BW.

CHAPTER 4*

Effects of backpack load on critical changes of trunk muscle activation and lumbar spine loading during walking in young male adults

* This chapter has been published by the author of this thesis.
Li, S. S. W., & Chow, D. H. K. (2018). Effects of backpack load on critical changes of trunk muscle activation and lumbar spine loading during walking. Ergonomics, 61(4), 553-565.

Critical changes featuring the disproportionality of increases in trunk muscle activation and lumbar joint loading while walking with backpack carriage in light and heavy weight may reveal the load-bearing strategy of the lumbar spine. This was evaluated using an integrated approach equipped with a wireless surface electromyography, a force platform, and a motion analysis system to measure the trunk muscle EMG amplitudes and lumbar joint component forces. This part of the study hypothesized that a multi-objective goal programming model could be identified and used to predict the most critical changes in trunk muscle activation (in terms of peak EMG amplitude) and lumbar spine loading (in terms of peak lumbosacral joint force) while walking with no-load and a backpack loaded up to 20% of BW.

4.1 Background

Walking is a usual human activity and considered to be harmless to healthy individuals, and even beneficial to low back pain patients (Nuttter 1988). Walking with backpack carriage is a typical daily activity for many working adults, college students, adolescents, and schoolchildren. However, walking with a heavy backpack has been identified as an ergonomic risk factor for low back pain (Heuscher et al. 2010).

Walking induces cyclic stress on the spine, and walking while carrying a loaded backpack further increases the mechanical loading on the lumbar spine. The additional spinal stress may be due to the change in trunk and pelvis coordination variability (Yen et al. 2012), non-synchronized movements of trunk and backpack (Pierrynowski et al. 1981), as well as increases in flexion (Singh and Koh 2009), trunk stiffness (Holt et al. 2003), and vertical ground reaction forces transmitted through the lower-body-linked segments up to the lower spinal levels (Birrell et al. 2007).

Numerous studies on lumbar spine loading during walking (Callaghan et al. 1999; Cheng et al. 1998; Goh et al. 1998; Khoo et al. 1995; Shojaei et al. 2016) have concluded that the peak compression force generated in the lumbar spine is between 143% and 309% (mean: around 185%) of BW, i.e. in absolute terms, between 1009 N and 2179 N with mean approximately 1305 N. Only one of these studies evaluated the impact of different backpack loads on spinal load. Walking with a backpack load at 15% or 30% of BW increased the peak compression force by 26% and 64% of BW, which were 11% and 34% of BW greater than the backpack weight, respectively (Goh et al. 1998). The disproportionality of increases in joint loading between carrying a light and heavy weight backpack reveals critical changes in the load-bearing strategy of the lumbar spine. However, there is a lack of detailed evaluation in the disproportionality of or critical changes in the increases in mechanical joint loading between backpack loads of 0%–15% and 15%–30% of BW.

The muscle and joint contact forces required to counterbalance the resultant moment of the joint are the two dominant components of joint load during walking. The joint compression contact force at lower spinal level is, on average, approximately 578 N (82% of BW) (Callaghan et al. 1999; Hendershot and Wolf 2014). However, the compression force acting at the joint due to trunk muscles could be as high as 1545 N (219% of BW) (Callaghan et al 1999). Backpack carriage at 10% of BW significantly increased abdominal muscle activation (Devroey et al. 2007) and might generate excessive mechanical loading at the lumbar joint and induce low back pain during prolonged walking (Knapik et al. 1996). However, in elders with spinal deformities, carrying a lightweight backpack significantly decreased back muscle activation and reduced their back pain (Ishida et al. 2008; Lai et al. 2011). Even in healthy individuals, there may be an indication of a slight decrease in overall lumbar spine loading from lightweight backpack walking.

Carrying a loaded backpack was considered an external perturbation on the spinal column during walking. Based on the weight of backpack, various LBSs were executed at the lumbar joint through trunk muscle activation patterns. While the heaviness of a backpack increases, different load-bearing strategies were catered at the lumbar joint through trunk muscle activation patterns. This part of the study considered the minimum or maximum rate of change in muscle activation and joint loading as the critical event in lumbar spine strategy to allocate the minimum or maximum muscle activation and joint force per unit increase in backpack load, and evaluated the impacts of backpack weight on critical changes in trunk muscle activation and lumbar spine loading while walking. The hypothesis of this part of the study was that a multi-objective goal programming approach could be identified and applied to determine the most critical changes in trunk muscle activation (in terms of peak EMG amplitude) and lumbar spine loading (in terms of peak lumbosacral joint force) while walking with no-load (0% of BW) versus loaded with a backpack of up to 20% of BW.

4.2 Methods

4.2.1 Participants

A convenience sample of 10 healthy male college students was recruited from a local college (Table 4.1). The sample size was estimated with reference to previous similar studies in young adults (Goh et al. 1998; Majumdar et al. 2010; Majumdar et al. 2013) and was justified by post-test observation of statistical power of 0.8 at significant level, p=0.05. Participants were screened and those with known history of any spinal deformity or low back pain were excluded. Their mean height, body weight, and age were 1.76 (SD 0.04) m, 71.3 (SD 8.9) kg, and 23.7 (SD 3.0) years, respectively. Ethical approval from the Human Research Ethics Committee and written informed consent from all participants were obtained prior to the experimentation.

Table 4.1 Characteristics of the ten healthy male Participants.

Participant	Height (m)	Weight (kg)	Age (year)
1	1.76	69.3	29
2	1.75	53.7	25
3	1.78	78.6	23
4	1.8	79.4	24
5	1.81	81	19
6	1.76	74.2	22
7	1.69	68.2	25
8	1.76	73.6	27
9	1.69	75.2	23
10	1.75	59.3	20
Mean	**1.76**	**71.3**	**23.7**
SD	**0.04**	**8.9**	**3.0**

4.2.2 *Experimental design*

A randomized repeated measure design was adopted in this part of the study. Independent variables were the backpack load (0% for no-load, and 5%, 10%, 15%, and 20% of BW for loaded conditions), and the side effect (left and right strides), where applicable. Dependent variables were (a) spatial–temporal parameters including cadence, walking speed, single-support duration, double-support duration, step length, step time, stride length, and stride time; (b) ground reaction forces including the first vertical peak, vertical trough, second vertical peak, as well as peak of anterior, posterior, and mediolateral shears; (c) peak electromyography (EMG) amplitudes of six bilateral pairs of trunk muscles including the rectus abdominis, external oblique, internal oblique, latissimus dorsi, thoracic erector spinae, and lumbar erector spinae; and (d) peak forces at the lumbosacral joint including compression, anterior, posterior, and mediolateral shears.

4.2.3 *Instrumentation*

The lower body segments were identified using 12-mm diameter reflective markers and their motions were captured by eight Oqus 700+ cameras (Qualisys, Gothenburg, Sweden). Trunk muscle activities were collected using a wireless surface EMG system (bandwidth CMMR > 80 dB at 60 Hz, bandwidth 20–450 Hz, gain 1000, Trigno, Delsys, Boston, MA, USA) at sampling rates of 100, 2000, and 2000 Hz, respectively. Ground reaction forces were recorded by three force platforms (Model 4060-10, Bertec Corporation, Columbus, OH, USA).

All data were acquired synchronously with a trigger pulse device (Delsys Trigger Module, Boston, USA) using the Qualisys Tracking Manager software (Qualisys, Gothenburg, Sweden). This part of the study adopted an internally reconstructed double-strap backpack used in a previous study (Li and Chow 2016) (Figure 3.1 in Chapter 3). The cameras were positioned so that the view of reflective markers, especially those affixed to the lower back, were not blocked by the backpack.

4.2.4 Data collection

Twenty anthropometric measurements (body mass, anterior superior iliac spine breadth, left and right thigh length, mid-thigh circumference, shank length and circumference, knee diameter, foot length and breadth, and malleolus height and width) were recorded for estimating the parameters of body segments (Vaughan et al. 1999).

Fifteen dynamic anatomical markers (second metatarsal heads, heels, lateral malleoli, tibia wands, lateral femoral epicondyles, femoral wands, and anterior superior iliac spines and sacrum) (Vaughan et al. 1999) and two static anatomical markers (posterior superior iliac spines) were affixed to the respective body landmarks (Figure 4.1).

The skin around the 12 identified placements of surface electrodes was cleaned using alcohol to remove dead skin, oil, and dirt, as well as shaved if necessary to reduce resistance during the EMG acquisitions. Then, 12 EMG electrodes were attached to each subject on these identified surfaces of the trunk muscles (McGill 1992): lumbar erector spinae (3 cm lateral to the L3 spinous process), thoracic erector spinae (5 cm lateral to the T9 spinous process), latissimus dorsi (lateral to T9 over the muscle belly), internal oblique (below the external oblique electrodes and just superior to the inguinal ligament), external oblique (approximately 15 cm lateral to the umbilicus), and rectus abdominis (3 cm lateral to the umbilicus) (Figure 4.1). The locations and arrangements of these six pairs of muscles have been validated to minimize the signal crosstalk between electrode pairs during bending and twisting tasks, and be the best representation of differential muscle activity (Lafortune et al., 1988).

Figure 4.1 Front and back views of the placements of surface EMG electrodes and anatomical markers.

Each participant was asked to maintain a relaxed barefoot upright stance on the force platforms for 3 sec of static trial capturing and then instructed to stand at the starting position of the walkway for 5 sec prior to performing the walking trials at his preferred speed along a 10-m walkway (embedded with 3 force platforms) under the no-load (0% of BW) and four loaded conditions (5%, 10%, 15%, and 20% of BW). The center of gravity of backpack under the loaded conditions was identified using a horizontal belt adjusting to the T12 spinal level by lengthening or shortening the shoulder straps (Li and Chow 2016). The sequence of loading conditions was randomized. Three successful trials were recorded for each loading condition. Prior to the static and walking trials, individual participant was asked to do four maximum voluntary contraction (MVC) activities: shoulder extensions, trunk extensions, back lift pull, and bent sit up (Gagnon et al. 2001). Each MVC trial was recorded for 3 sec. Subjects rested for 1–2 min between two consecutive experimental trials and 5 min after finishing an entire set of walking or MVC tasks.

4.2.5 Data processing

Raw marker and force platform data were filtered using a bidirectional fourth-order Butterworth low pass filter at cutoff frequencies of 6 and 25 Hz, respectively (Hendershot and Wolf 2014). Raw EMG data were full-wave rectified and processed using dual-pass, fourth-order Butterworth low pass filter at a cutoff frequency of 2.5 Hz (Pakzad et al. 2016). All filtering processes were implemented using the MATLAB 2015b software (The MathWorks, Inc., Natick, MA, USA).

The period of a complete gait cycle of each walking trial was identified between two consecutive left foot initial contacts on the ground. These two critical events as well as left and right foot toe off were determined from the vertical ground reaction force recorded by the force platforms using a threshold of 20 N (Zeni Jr. et al. 2008) as well as the markers affixed to the heels and second metatarsal heads (Cappellini et al., 2006). The spatial-temporal parameters were measured based on the chronological events within an identified gait cycle: left foot initial contact, right foot toe off, right foot initial contact, left foot toe off, and left foot initial contact (Ayyappa, 1997). The gait cycle of each walking trial was time-normalized to 101 points for further data processing. The EMG signals of the walking trials were normalized using the respective EMGs at MVCs. A previous study demonstrated that the EMG amplitudes of maximum muscle contraction levels was not always reliable while on the other hand the peak level determined to be higher no more than 10% of an identified cycle time would provide more relevant information (Jonsson 1982). Thus, this part of the study catered for the peak EMG amplitudes at the 90 percentile of the normalized EMG amplitudes.

The International Society of Biomechanics recommendations for standardization in the reporting of kinematic data (Wu and Cavanagh 1995) and definitions of the joint coordinate system (Wu et al. 2002) were adopted to define the global and anatomical coordinate systems. The lumbosacral joint center was estimated by the bony landmarks of anterior and posterior superior iliac spines, and the sacrum (Reed et al. 1999). The baseline lumbosacral disc angle was based on the previous literature (McGill and Norman 1986). A lower body inverse dynamic model was used to calculate the lumbosacral joint reaction force and moment (Plamondon et al. 1996). The lower body model covered seven segments including feet, shanks, thighs, and pelvis. Reflective markers were attached to each segment for determining its linear and angular accelerations. Inverse dynamics was applied to determine the linear and angular inertial effects of each segment. Together with the measured ground reaction forces, the resultant force and moment at the lumbosacral joint were determined. The moment was then partitioned among the six bilateral pairs of trunk muscles by an upper body trunk musculoskeletal architecture providing the anthropometric data of muscle cross-sectional areas, lever arms, and lines of action (McGill and Norman 1986; Gagnon et al. 2001). An EMG-assisted optimization model developed by Cholewicki et al. (1995) was adopted, except that the passive muscle forces were neglected because of its low magnitude in upright trunk posture (Cholewicki et al., 1995) and the lower bound of muscle gains, adjusting the initial muscle force estimates to perfectly counter balance the moments induced in the lumbar spine, were set at 0.5 (instead of zero) for all muscles for ensuring adequate prior EMG-based muscle force estimates for all trunk muscles (Gagnon et al., 2001). The objective of the optimization approach was to determine the least possible adjustments to the initial muscle force estimations while satisfying the perfect moment equilibrium at the lumbosacral joint.

Trend lines were used to fit the scatter plots of the overall means of peak EMG amplitudes and joint forces with a significant backpack load effect. An appropriate regression model for overall means was identified. The derivative functions of regressions representing the rates of changes in peak EMG amplitudes and joint forces were obtained. The maximum or minimum rates of changes in peak EMG amplitudes and joint forces were determined and defined as critical changes in individual muscle and joint component force. The maximum or minimum rates of changes in peak EMG amplitudes and joint forces could appear at different points along the continuous backpack-loading spectrum between 0% and 20% of BW. The most critical rates of changes in peak EMG amplitudes and joint forces were determined by locating the position with the least deviation from the maximum or minimum rates of changes. A multi-objective GP model (Li and Chow 2016) was adopted to determine these positions.

4.2.6 Statistical analysis

The averages of three successful trials of each dependent variable for each loading condition were used as input data. A one-way or two-way MANOVA was used to analyze the significance of backpack load, side, or backpack load and side interaction on the four groups of dependent variables: spatial-temporal parameter, ground reaction force, peak EMG amplitude, and peak joint force. A repeated measure ANOVA was then used to analyze those groups of dependent variables with significant effects (SPSS version 21.0, IBM Inc., Chicago, IL, USA). Statistical significance and power were set at $p=0.05$ and $p=0.8$, respectively. Post-hoc comparisons were made based on the least significant difference (LSD) criterion.

4.3 Results

The spatial-temporal parameters, ground reaction forces, peak EMG amplitudes, and lumbosacral joint forces under various loading conditions for the ten participants were obtained (Appendices 4.1, 4.2, 4.3, 4.4). The pooled means of all dependent variables were figured out (Table 4.2).

The results of two-way MANOVA tests (Table 4.3a) showed that there were no significant side or backpack load and side interaction effect on all the three groups of dependent variables: spatial-temporal parameters including single-support duration, step length, and step time ($p=0.441$ and 0.607, respectively), ground reaction forces ($p=0.619$ and 1.000, respectively), and peak EMG amplitudes ($p=0.150$ and 1.000, respectively). The backpack load effect was not significant on the group of spatial-temporal parameters ($p=0.945$), but significant on both groups of ground reaction forces ($p<0.001$) and peak EMG amplitudes ($p=0.007$).

The results of one-way MANOVA tests (Table 4.3b) showed that the backpack load effect was significant on the peak joint forces ($p<0.001$), but not significant on the spatial-temporal parameters including cadence, walking speed, double-support duration, stride length, and stride time ($p=0.983$).

The significant backpack load effects on the ground reaction forces, peak EMG amplitudes, and peak joint forces were further analyzed by repeated measure ANOVA (Table 4.3c). The effect of backpack load was significant on all dependent variables ($p<0.05$), except that the changes in peak EMG amplitudes of latissimus dorsi ($p=0.233$) and thoracic erector spinae ($p=0.940$) were not significant.

Table 4.2 Pooled means of all response variables from all participants. The figures shown inside the bracket were standard errors of the means.

Backpack load		0% BW	5% BW	10% BW	15% BW	20% BW
Spatial-temporal parameter	Cadence (steps/min)	120 (2.6)	120 (1.8)	118 (2.0)	118 (2.5)	117 (2.4)
	Walking speed (m/s)	1.29 (0.03)	1.30 (0.02)	1.27 (0.03)	1.27 (0.03)	1.25 (0.03)
	Double support duration (% GC)	10.6 (0.4)	10.6 (0.3)	11.4 (0.4)	11.6 (0.3)	11.3 (0.6)
	Single support duration (% GC)	39.4 (0.4)	39.4 (0.5)	38.6 (0.5)	38.4 (0.3)	38.7(0.7)
	Step length (m)	0.64 (0.01)	0.65 (0.01)	0.64 (0.01)	0.65 (0.01)	0.64 (0.01)
	Step time (s)	0.50 (0.01)	0.50 (0.01)	0.51 (0.01)	0.51 (0.01)	0.51 (0.01)
	Stride length (m)	1.29 (0.02)	1.30 (0.02)	1.29 (0.02)	1.30 (0.02)	1.28 (0.02)
	Stride time (s)	1.00 (0.02)	1.00 (0.01)	1.02 (0.02)	1.02 (0.02)	1.03 (0.02)
Ground reaction force (% BW)	First peak of vertical	112 (1.7)	118 (1.5)	121 (1.5)	125 (2.1)	130 (1.7)
	Second peak of vertical	109 (1.1)	115 (1.4)	121 (1.2)	125 (1.2)	130 (1.1)
	Trough of vertical	72 (1.1)	73 (1.9)	79 (1.4)	84 (1.0)	87 (1.1)
	Peak of anterior shear	23.3 (0.8)	24.5 (0.8)	25.1 (0.8)	26.0 (0.8)	27.3 (0.8)
	Peak of posterior shear	20.2 (0.9)	21.7 (0.7)	22.0 (0.7)	23.4 (0.8)	24.2 (0.8)
	Peak of mediolateral shear	7.5 (0.4)	7.6 (0.3)	8.0 (0.3)	8.4 (0.3)	8.6 (0.4)
Peak EMG amplitude (% EMG_{MVC})	Rectus abdominis	3.2 (0.4)	3.5 (0.4)	4.4 (0.6)	6.0 (0.9)	7.0 (07)
	External oblique	6.5 (0.9)	7.1 (1.1)	8.1 (1.2)	10.1 (1.4)	8.7 (1.1)
	Internal oblique	8.4 (0.9)	8.7 (1.1)	9.6 (1.1)	11.3 (1.0)	11.1 (1.2)
	Latissimus dorsi	4.3 (0.5)	3.8 (0.5)	4.6 (0.7)	3.7 (0.4)	4.2 (0.5)
	Thoracic erector spinae	7.0 (0.6)	6.6 (0.6)	6.9 (0.9)	6.5 (07)	6.9 (0.5)
	Lumbar erector spinae	8.6 (0.5)	8.0 (0.6)	7.0 (0.6)	6.7 (0.6)	7.4 (0.5)
Peak lumbosacral joint force (% BW)	Compression	195 (5.2)	190 (4.9)	210 (5.3)	234 (6.1)	255 (6.5)
	Anterior shear	25.0 (1.7)	25.7 (2.3)	26.5 (2.7)	28.3 (2.4)	33.4 (2.7)
	Posterior shear	19.6 (2.3)	20.9 (2.3)	24.5 (2.7)	25.0 (3.4)	26.4 (2.7)
	Mediolateral shear	15.1 (0.8)	15.5 (0.8)	17.8 (1.3)	18.6 (1.4)	20.0 (1.6)
BW = body weight, GC = gait cycle						

Table 4.3 Summary of the statistical results of the backpack load, side and their interaction. (a) Two-way MANOVA, (b) One-way ANOVA, and (c) Repeated measure ANOVA.

		p-value			
(a) Two-way MANOVA		**Backpack load**	**Side**	**Backpack load and side interaction**	**Statistical power**
Spatial-temporal parameter (single-support duration, step length and time)		0.945	0.441	0.607	-
Ground reaction force		< 0.001	0.619	1.000	1.000
Peak EMG amplitude		0.007	0.150	1.000	0.976
(b) One-way MANOVA					
Spatial-temporal parameter (cadence, walking speed, double support-duration, stride length and time)		0.983	-	-	-
Peak joint force		< 0.001	-	-	0.998
(c) Repeated measure ANOVA with backpack load as within-subject factor					
Ground reaction force	First peak of vertical	< 0.001	-	-	1.000
	Trough of vertical	< 0.001	-	-	1.000
	Second peak of vertical	< 0.001	-	-	1.000
	Peak anterior shear	< 0.001	-	-	1.000
	Peak posterior shear	<0.001	-	-	0.999
	Peak medial-lateral shear	0.004	-	-	0.916
Peak EMG amplitude	Rectus abdominis	< 0.001	-	-	0.986
	External oblique	< 0.001	-	-	0.990
	Internal oblique	0.008	-	-	0.870
	Latissimus dorsi	0.233	-	-	-
	Thoracic Erector spinae	0.940	-	-	-
	Lumbar Erector spinae	0.001	-	-	0.965
Peak lumbosacral joint force	Compression	< 0.001	-	-	1.000
	Anterior shear	< 0.001	-	-	0.995
	Posterior shear	0.002	-	-	0.953
	Medial-lateral shear	0.003	-	-	0.937

4.3.1 Effect of loading on spatial–temporal parameters

Increases in backpack load resulted in no significant changes in cadence, walking speed, double support duration, single support duration, step length, step time, stride length, and stride time (Table 4.3a,b). The approximate double- and single-support durations (11% and 39% of gait cycle, respectively), step length (0.65 m) and time (0.5 s), and stride length (1.3 m) and time (1.0 s) were consistent across different loadings. When comparing between the no-load and loaded conditions, the cadence and walking speed slightly and gradually decreased from 120 steps to a minimum of 117 steps and from 1.29 m/s to a minimum of 1.25 m/s, respectively (Table 4.2).

4.3.2 Effect of loading on ground reaction forces

Increases in backpack load significantly increased the ground reaction forces including the first peak of vertical ($p<0.001$), trough of vertical ($p<0.001$), second peak of vertical ($p<0.001$) as well as peaks of anterior shear ($p<0.001$), posterior shear ($p<0.001$), and mediolateral shear ($p=0.004$) (Table 4.3c). Significant changes occurred when the backpack was loaded at and beyond 5%, 10%, 5%, 5%, 10% and 15% of BW, respectively (Figure 4.2).

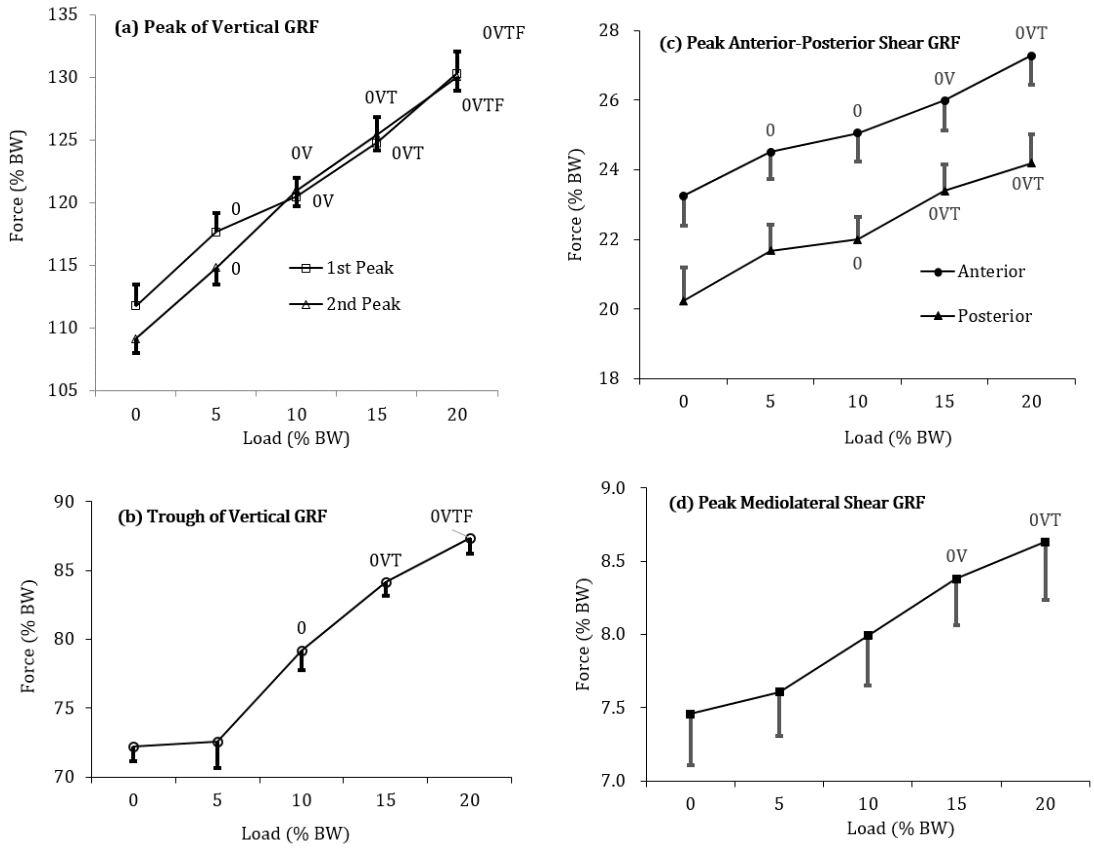


Figure 4.2 Pooled means of the ground reaction forces (GRFs) during walking under various loading conditions. One-sided vertical bars indicate standard error of the mean (SEM). Data labels 0, V, T, and F indicate significant differences ($p < 0.05$) at 0%, 5%, 10%, and 15% of BW respectively.

4.3.3 Effect of loading on peak EMG amplitudes

The pooled means of peak EMG amplitudes of the latissimus dorsi and thoracic erector spinae were slightly decreased when comparing between the no-load and loaded conditions (Table 4.2). However, the effects were not significant (Table 4.3c). The increase in backpack load from the baseline (no-load) condition significantly increased the peak EMG amplitudes of the rectus abdominis and external and internal oblique and significantly reduced the peak EMG amplitudes of the lumbar erector spinae at loadings of ≥10%, 10%, 15% and 10% of BW (Figure 4.3). With an increase in backpack load, the rate of changes in the peak EMG amplitudes of the rectus abdominis, external and internal oblique, and lumbar erector spinae changed from the baseline condition until they reached their respective critical loads at 12.1%, 8.8%, 10.0%, and 7.1% of BW, and then reversed towards reduction or increment with further escalation in backpack load (Figure 4.3).

4.3.4 Effect of backpack load on peak joint forces

Increases in backpack load from the baseline (no-load) condition significantly increased the peak compression and anterior, posterior, and mediolateral shear joint forces at loadings of ≥10%, 20%, 10% and 15% of BW (Figure 4.4). All three peak shear forces strictly increased with increases in backpack load. However, the minimum peak compression force was observed at a backpack load of 3.2% of BW. With an increase in backpack load, the rates of changes in the peak compression and posterior and mediolateral shear joint forces changed from the baseline condition until they reached their respective critical loads at 12.8%, 7.7%, and 11.1% of BW. Then, they reversed towards reduction with further escalation in backpack load (Figure 4.4a,c,d). By contrast, the rate of change of the peak anterior shear joint force decreased at the beginning of the loading until the critical load was reached at 5.5% BW, before reversing to increment with further increases in backpack load (Figure 4.4b).

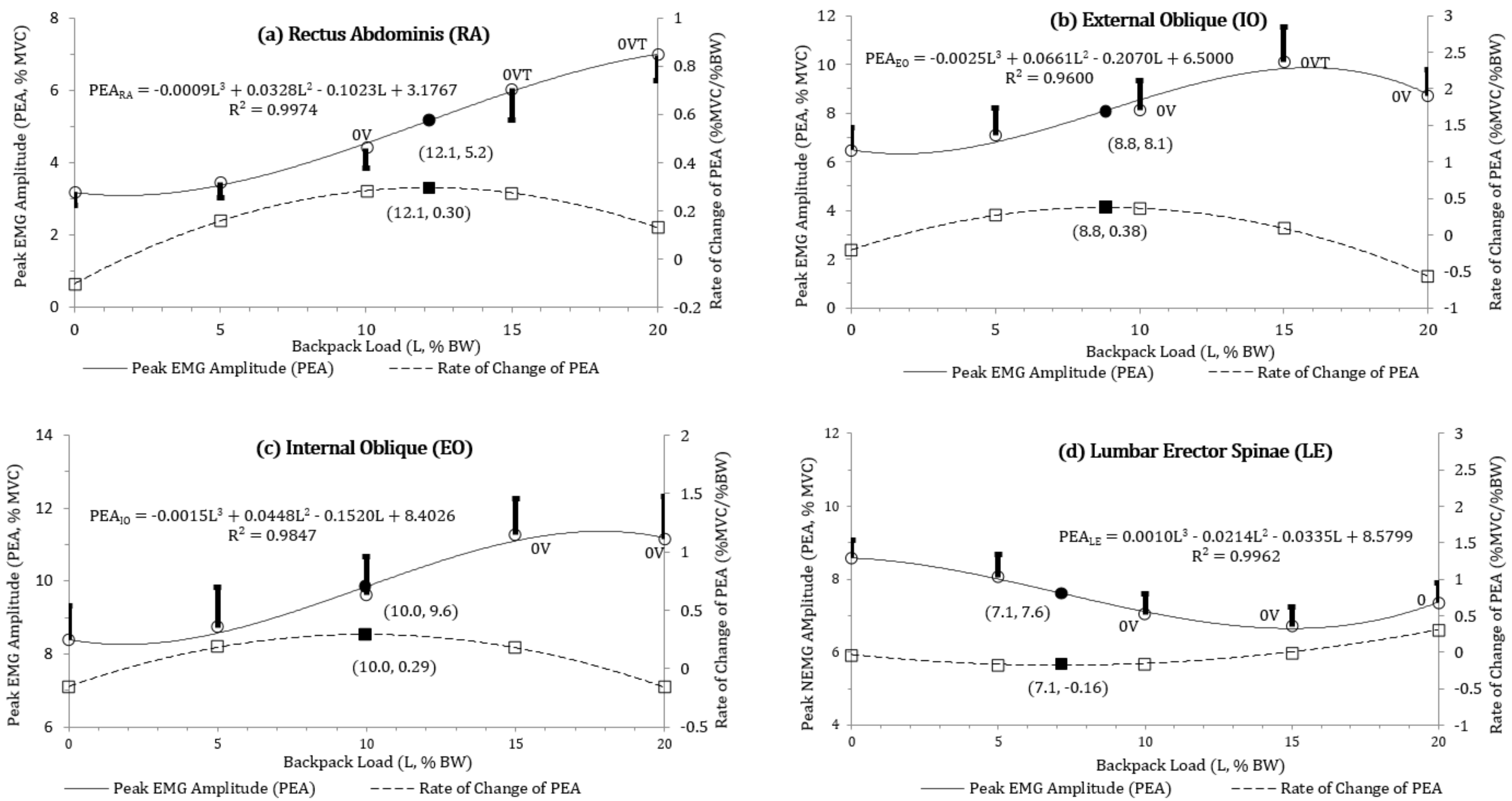


Figure 4.3 Pooled means of peak EMG amplitudes (PEA) and rate of change of PEA (RCPEA) of (a) rectus abdominis, (b) external oblique, (c) internal oblique, and (d) lumbar erector spinae during level walking for backpack loads of 0%, 5%, 10%, 15%, and 20% of BW. One-sided vertical bars indicate standard error of the mean (SEM). Vertical scales are not standardized. Data labels 0, V, T, and F indicate significant differences at 0%, 5%, 10%, and 15% of BW, respectively. Third-order polynomial regression lines and their derivative functions are also shown. Black marks represent the critical positions of PEA (circles) and RCPEA (squares).

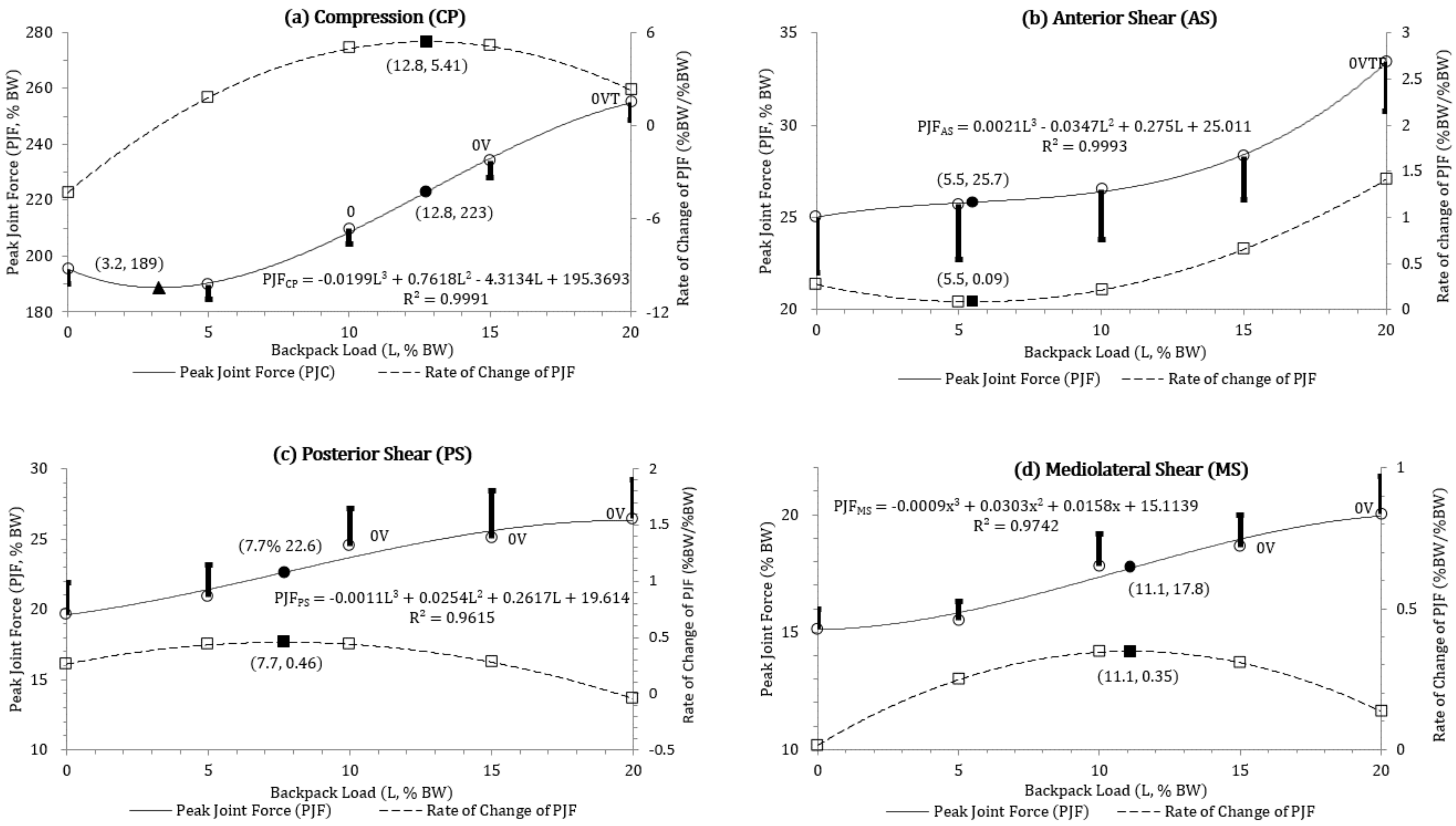


Figure 4.4. Pooled means of peak joint force (PJF) and rate of change of PJF (RCPJF) of (a) compression (CP), (b) anterior shear (AS), (c) posterior shear (PS), and (d) mediolateral shear (MS) at the lumbosacral spinal level during level walking for backpack loads of 0%, 5%, 10%, 15%, and 20% of BW. One-sided vertical bars indicate standard error of the mean (SEM). Vertical scales are not standardized. Data labels 0, V, T, and F indicate significant differences ($p < 0.05$) at 0%, 5%, 10%, and 15% of BW, respectively. Third-order polynomial regressions and their derivative functions are also shown. Black marks represent the critical positions of PJF (circles) and RCPJF (squares), and the minimum position of PJFCP (triangle). at 0%, 5%, 10%, and 15% of BW, respectively. Third-order polynomial regression lines and their derivative functions are also shown. Black marks represent the critical positions of PEA (circles) and RCPEA (squares).

4.3.5 Rates of change in muscle activities and joint forces

Individual backpack loads with respect to critical changes in peak EMG amplitudes and joint forces were 12.1% (rectus abdominis), 8.8% (external oblique), 10.0% (internal oblique), 7.1% (lumbar erector spinae), 12.8% (compression), 5.5% (anterior shear), 7.7% (posterior shear), and 11.1% (mediolateral shear) of BW. The Chebyshev variant goal programming model (Li and Chow 2016) applied for both the rate of change of peak EMG amplitudes and joint forces was formulated in accordance with the respective cost functions, constraints, and boundary and non-negativity conditions (Table 4.4). The most critical backpack loads with respect to the overall minimum deviation from individual critical changes in peak EMG amplitudes and joint forces were 10.32% and 10.48% of BW, respectively.

4.4 Discussion

4.4.1 Spatial–temporal parameters

Although increasing backpack load up to a maximum of 20% of BW had no significant effects on any spatial–temporal parameter in this part of the study, a trend of a gradual increase in the stride duration and a decrease in both cadence and walking speed was observed. The findings of the significant effects of backpack load on spatial–temporal parameters were not consistent between adolescent and young adult age groups. The results of this part of the study were

Table 4.4 Chebyshev variant goal programming models for the rate of change of (a) peak EMG amplitudes and (b) peak joint forces: formulation with respect to backpack load and in accordance with the respective cost function, constraints, boundary, and non-negativity condition.

(a) Rate of change peak EMG amplitude	
Cost Function	k
Constraints	-0.0027*L*L + 0.0656*L $-D_{RA}^{+} + D_{RA}^{-}$ = 0.3985 -0.0075*L*L + 0.1322*L $- D_{EO}^{+} + D_{EO}^{-}$ = 0.5826 -0.0045*L*L + 0.0896*L $- D_{IO}^{+} + D_{IO}^{-}$ = 0.4460 0.0030*L*L - 0.0428*L $- D_{LE}^{+} + D_{LE}^{-}$ = 0.1527 3.3766* $D_{RA}^{-} \leq k$ 1.7798*$D_{EO}^{-} \leq k$ 3.4013*$D_{IO}^{-} \leq k$ 5.3719*$D_{LE}^{+} \leq k$
Boundaries	$L \geq 0, L \leq 20$
Non-negativity	$D_{RA}^{-}, D_{RA}^{+}, D_{EO}^{-}, D_{EO}^{+}, D_{IO}^{-}, D_{IO}^{+}, D_{LE}^{-}, D_{LE}^{+} \geq 0$
Solution	L = 10.32
Abbreviations	D_i^{+}: positive deviation from critical change of muscle i D_i^{-}: negative deviation from critical change of muscle i i = RA (rectus abdominis), IO (internal oblique), EO (external oblique), LE (lumbar erector spinae)

(b) Rate of change of peak joint force	
Cost Function	k
Constraints	-0.0597*L*L + 1.5236*L $- D_{CP}^{+} + D_{CP}^{-}$= 9.7209 0.0063*L*L - 0.0692*L $- D_{AS}^{+} + D_{AS}^{-}$= -0.1900 -0.0033*L*L + 0.0506*L $- D_{PS}^{+} + D_{PS}^{-}$ = 0.1940 -0.0027*L*L + 0.0600*L $- D_{LS}^{+} + D_{LS}^{-}$= 0.3333 0.1849* $D_{CP}^{-} \leq k$ 11.8239*$D_{AS}^{-} \leq k$ 2.1893*$D_{PS}^{-} \leq k$ 2.8667*, $D_{LS}^{-} \leq k$
Boundaries	$L \geq 0, L \leq 20$
Non-negativity	$D_{CP}^{-}, D_{CP}^{+}, D_{AS}^{-}, D_{AS}^{+}, D_{PS}^{-}, D_{PS}^{+}, D_{LS}^{-}, D_{LS}^{+} \geq 0$
Solution	L = 10.48
Abbreviations	D_i^{+}: positive deviation from critical change of force i D_i^{-}: negative deviation from critical change of force i i = CP (compression), AS (anterior shear), PS (posterior shear), MS (mediolateral shear)

comparable to the findings in young adults and the trend of slight changes in spatial–temporal parameters was largely similar to the findings in adolescents (Chow et al. 2005a; Devroey et al. 2007; Goh et al. 1998; Majumdar et al. 2013). Studies have shown that increasing the walking speed increased the peak compression and anterior–posterior shear forces (Callaghan et al. 1999; Cheng et al. 1998) and the peak lateral–medial shear force (Cheng et al. 1998) at lower lumbar spine joints. The possible confounding effect of walking speed on joint loads was controlled and minimized by the non-significant changes in walking speed under the various backpack-loading conditions of this part of the study.

4.4.2 Ground reaction forces

The general statistical finding of this part of the study was that the heavier the backpack load, the higher the components of ground reaction forces during level walking. The significant effects of backpack load on ground reaction forces were largely consistent with those reported in previous studies (Chow et al. 2005a; Majumdar et al. 2013). Changes in the load transfer of ground reaction forces transmitted through the lower-body-linked segments to the lower spinal levels were largely proportional to the system weight (body plus backpack) and similar between light and heavy backpack carriage weight. Critical changes in ground reaction forces were not observed along the backpack-loading spectrum between 0% and 20% of BW.

4.4.3 Peak EMG amplitudes

In general, the peak activities of flexor muscles (rectus abdominis and external and internal oblique) increased and those of the extensor muscle (lumbar erector spinae) decreased with an increase in backpack load. However, the peak muscle activity of the external oblique decreased and that of the lumbar erector spinae increased when the backpack load was increased from 15% to 20% of BW. Devorey et al. (2007) corroborates the significant changes in the activities of the rectus abdominis, external oblique, and lumbar erector spinae during the gait cycle under loads of up to 15% of BW. During the strike of the gait, the mean linear envelope of the normalized EMGs of the rectus abdominis, external oblique, and lumbar erector spinae decreased with increased backpack loads of up to 15% of BW. The results of this part of the study were largely comparable to these findings. The abdominal and back muscles do not exhibit a cocontraction strategy with an increase in backpack load. It is noteworthy that the perturbation of the external backpack load is passively handled, indicating the critical response of the musculoskeletal structure of the trunk along the backpack loading spectrum between 0% and 20% of BW.

4.4.4 Peak joint forces

Walking with backpack carriage at an average speed of 1.29 m/s induced overall mean peak compression of 195% of BW (1364N) and anterior, posterior, and mediolateral shear forces of 25% (175 N), 18% (140N), and 15% of BW, (105N), respectively. These results are within the range of findings reported in previous studies (Callaghan et al. 1999; Cheng et al. 1998; Goh et al. 1998; Khoo et al. 1995; Shojaei et al. 2016). Of these five studies, only Goh et al. (1998) investigated the effects of varying backpack loads (0%, 15%, and 30% of BW) on peak joint forces. The peak compression and resultant shear forces at the lumbosacral joint significantly increased by 26.7% and 25.0%, respectively, when the backpack load was increased from 0% to 15% of BW, and increased by 29.5% and 28.6%, respectively, when the backpack load was increased from 15% to 30% of BW. In this part of the study, the percentages of increase in peak compression, anterior, posterior, and mediolateral shear forces between 0% and 15% BW, and between 15% and 20% BW were 19.9 and 8.9 % (272 and 146 N), 13.2 and 18.0 % (23.2 and 35.7 N), 27.6 and 5.6 % (37.8 and 9.9 N), and 23.1 and 7.6 % (24.5 and 9.9 N), respectively. The higher percentage of an increase in the peak compression force at backpack load of 15% of BW reported in the study of Goh et al. (1998) might be attributable to the increase in walking speed at 1.31 m/s when compared with 1.22 m/s in the unloaded condition. The disproportionality of increases in joint loading between light and heavy backpack carriage weight revealed critical changes in the LBS of the lumbar spine along the backpack-loading spectrum between 0% and 20% of BW.

The changes for backpack loads from 0% to 20% BW (0 to 140 N), on average, increased compression force by 420 N (from 195% BW—1364 N to 255% BW—1784 N) and anterior shear force by 59 N (from 25.0% BW—175 N to 33.4% BW—234 N) (Figure 4.4a,d). The change of compression force was superior to the increase in backpack load, but the increase in shear force was so small that measurement was prone to error. Arjmand et al. (2009) evaluated two biomechanical models (single-joint EMG-assisted optimization and multi-joint kinematics-driven finite element models) under various static lifting activities. They reported that both models employed local lumbar and global trunk muscles in various proportions, and predicted comparable compression forces but significantly different shear forces (more than 100 N). The accuracy of indirect measurements of spinal load remains unknown, but models are prone to errors particularly in the measurements of shear forces.

4.4.5 Changes in muscle activities and joint forces

The external perturbation of backpack carriage adds both weight and mass moment of inertia to the body at the same time (Costello et al. 2012) and shifts the body's center of mass posteriorly (Goh et al. 1998). A forward lean of the trunk in compensation moves the body's center of mass anteriorly. The LBS at the lumbar spine is largely incorporated with the movement that is activated by the trunk muscles. Muscle forces induced at the lumbar spine could account for more than two-thirds of the peak compression during walking (Callaghan et al. 1999).

In this part of the study, carrying a lightweight backpack at 3.2% of BW may have reduced the peak lumbosacral compression force at approximately 3% of BW compared with no-load condition. As loading increased, the lumbosacral joint supported more weight, and the peak joint forces increased disproportionally until they reached the maximum backpack weight of 20% of BW. The rate of change of peak EMGA amplitudes and joint forces representing the load-bearing efficiency of the lumbar spine may reveal the incorporated mechanism between trunk muscle activation and lumbar spine loading strategies. While the rate of change of peak EMGA amplitudes and joint forces represent critical changes in individual trunk muscle activation and joint component force, the most critical rate of change of the peak EMGA amplitude and joint force representing overall trunk muscle activation and lumbar spine loading was found to be at a backpack load of approximately 10% of BW. The most critical backpack load at 10% of BW indicates the occurrence of a maneuver change in the LBS in the lumbosacral joint when the lack of trunk muscle co-contraction activation becomes prominent and significant.

The magnitudes and changes in peak trunk muscle activities as well as the changes in both joint and ground reaction shear forces measured or predicted for backpack loads (from 0% to 20% BW) in this part of the study were relatively small (Figures 4.3 and 4.4). Such changes may not be harmful to healthy young adults for a 10 m walking performed in this study when comparing with a prolonged, such as 5 km walking that may induce fatigue and back pain. Nevertheless, the load-bearing strategy of the lumbar spine required to maintain the spinal stability is a major concern for the external perturbation of the backpack loaded between 0% and 20% BW.

4.4.6 Load-bearing strategies

The compensation effect of muscle activities of the abdominal, oblique, and erector spinae muscle groups on the lumbosacral joint loads illustrated in Table 4.2 showed important indication of load-bearing strategy. The rate of change profiles of peak EMG amplitudes and joint component forces were partitioned along the continuum of the backpack load axis in accordance with their signs: positive (+) or negative (−) for increases or decreases in the peak EMG amplitude or joint component force, respectively (Table 4.5).

The profiles of peak EMG amplitudes were partitioned into five stages: 0–2%, 2–15%, 15–16%, 16–18%, and 18–20% of BW. The respective LBSs of the rectus abdominis, external oblique, internal oblique, and lumbar erector spinae were [−, −, −|−] when backpack load was increased from 0% of BW (baseline condition) to 2% of BW (Stage I); [+, +, +|−] when increased from 2% to 15% of BW (Stage II); [+, +, + |+] when increased from 15% to 16% of BW (Stage III); [+ , − , + |+] when increased from 16% to 18% of BW (Stage IV); and [+ , −, −|+] when increased from 18% to 20% of BW (Stage V).

In the beginning (Stage I), both abdominal and back muscles catered to a synchronized LBS (all negative changes) that might indicate a reduction of lower level lumbar joint load from a lightweight backpack. By the end (Stage V), the rectus abdominis and lumbar erector spinae maintained a synchronized LBS by reversing from a negative to a positive rate of change, whereas both external and internal oblique catered to a non-synchronized LBS that might prepare for heavier backpack carriage exceeding 20% of BW.

Table 4.5 Partitions of the rate of change profiles of trunk muscle activation and lumbosacral joint load along the continuum of the backpack load axis in accordance with their signs: positive (+) or negative (-) for increases or decreases in peak EMG amplitude or joint component force, respectively.

Peak trunk EMG amplitude		Backpack load (% BW)					Peak lumbosacral joint force		Backpack load (% BW)	
		0-2	2-15	15-16	16-18	18-20			0-3	3-20
Abdominal muscle	Rectus abdominis	-	+ [12.1]	+	+	+	Compression		-	+ [12.8]
	External oblique	-	+ [8.8]	+	-	-	Shear	Anterior	+	+ [5.5]
	Internal oblique	-	+ [10.0]	+	+	-		Posterior	+	+ [7.7]
Back muscle	Lumbar erector spinae	-	- [7.1]	+	+	+		Mediolateral	+	+ [11.1]
Most critical change of trunk muscle activation			10.32				Most critical change of lumbosacral joint force			10.48

- : negative rate of change

+ : positive rate of change

[ ] : critical change of individual peak trunk muscle EMG or joint component force

In Stage II (2% to 15% BW backpack load), the back (negative changes) and abdominal muscles (positive changes) catered to a non-synchronized LBS, indicating the regions where a critical change in the individual peak EMG amplitude occurred (12.1% for rectus abdominis, 8.8% for external oblique, 10.0% for internal oblique, and 7.1% for lumbar erector spinae), as well as the regions where most critical changes in the overall LBS occurred (10% of BW).

The profiles of peak joint forces were partitioned into two stages: 0–3% and 3–20% of BW. The respective joint forces of compression and anterior, posterior, and mediolateral shear forces were [− |+, +, +] when backpack load was increased from 0% of BW (baseline condition) to 3% of BW (Stage A) and [+|+, + , +] when the load was increased from 3% to 20% of BW (Stage B). In the beginning (Stage A) and at the end (Stage B), all shear force rates of change were positive, whereas the compression force changed from negative to positive. The negative rate of change of the compression force in Stage A might indicate a reduction in compression joint force for lightweight backpack carriage. This result was synchronized with the LBS of trunk muscle activation reduction with a lightweight backpack. In Stage B (3% to 20% of BW backpack load), compression and all shear forces catered a synchronized LBS, indicating the regions where critical changes in the individual component joint forces occurred (12.8% for compression shear, 5.5% for anterior shear, 7.7% for posterior shear, and 11.1% for mediolateral shear), as well as the regions where most critical changes in the overall LBS occurred (10% of BW).

With light backpack carriage weight (Stage I of trunk muscle activation and Stage A of lumbosacral joint loading), the negative changes in both abdominal and back muscles cause a slight reduction in compression but not shear forces. This may be due to the backward shift of the body's center of mass increasing the moment of inertia of the system (body plus backpack), demanding a slight deactivation of both abdominal and back muscles. The trunk muscles progress through four different LBSs (Stages II to V) with an increase in backpack load to up to 20% of BW, along with incorporation of increasing lumbosacral joint forces (Stage B). Transitioning from the critical changes in abdominal muscles in Stage II, all the trunk muscles in Stage III cater to a synchronized or co-contraction LBS that may cope with the demand for stability control in flexion–extension, lateral bending, and twisting of the trunk when more backpack weight is added during walking. However, the co-contraction LBS further increases the lumbosacral joint force. The reduction in the muscle activation of the two oblique muscles in Stages IV and V contribute to alleviate the strictly increasing lumbosacral joint load while continuous increases in both flexor and extensor muscles play a major role in the overall stability by controlling trunk rigidity. This LBS may be a preparation for heavier backpack carriage exceeding 20% of BW.

4.4.7 Limitations

This part of the study has some limitations. First, for the convenience of conducting the experiment by a male experimenter, all subjects were young adult men, and the findings may not be generalizable to other gender or age groups. Second, the EMG-assisted optimization biomechanical spine model adopted to predict the trunk muscle forces did not consider the modulation factors of muscle length and velocity during dynamic locomotion, which may have induced computation errors for determining lower level spinal loading. Moreover, the consideration of moment equilibrium at individual joint at the lower lumbar spine and introduction of parameters such as the gain factor during the optimization process may induce computational error which can reduce the predictive accuracy of lumbar spine loading (Arjmand et al., 2009; Dreischarf et al., 2016). Third, this part of the study positioned the center of gravity of the backpack carriage at the T12 level which might be different among users as setting of backpack at lower or higher vertical position can also vary trunk muscle activation (Devroey et al., 2007). Fourth, this part of the study aimed at investigating the short-term effect of backpack load carriage and the results may not be applicable to long-term carriage as trunk muscle activation pattern may vary with fatigue, which can have effects on lumbar joint loading. Fifth, small sample size of 10 participants and only 5 level of backpack loading conditions were tested in this study, the overfitting of current data to the proposed polynomial regression model might be a concern of generalizing the evidence of the conclusive results reported in this part of the study. Larger scale and more detail researches are recommended for future study on the effects of backpack load on loading bearing strategy in the spine or at the joints of lower limbs during walking.

4.5 Summary

This part of the study investigated the effects of backpack load on critical changes in trunk muscle activation (in terms of peak EMG amplitude) and lumbar spine loading (in terms of peak joint force) during walking. A predictive goal programming model was applied to determine these critical changes during walking between no-load condition (0% BW) and backpack load carriage of up to 20% of BW. Light backpack carriage weight at approximately 3% of BW may reduce the cyclic peak compression force at the lumbosacral joint during walking. The most critical changes in both peak EMG amplitudes and joint forces were found to be at backpack load of approximately 10% of BW. The most critical changes in trunk muscle activation may be a good indicator for evaluating the LBS in the lumbar spine during backpack carriage due to the lack of co-contraction strategy between the abdominal trunk and back muscles.

CHAPTER 5*

Refined EMG-assisted optimization approach under optimal boundary condition for predicting lumbar spine loading during walking with backpack carriage in young male adults

* This chapter has been prepared for publication by the author of this thesis.
Li, S. S. W., Zheng, Y. P., & Chow, D. H. K. Refined EMG-assisted optimization approach under optimal boundary condition for predicting lumbar spine loading during walking with backpack carriage. (under revision)

This part of the study developed a computational algorithm for refining an EMG-assisted optimization approach for predicting lumbar spine loading while walking with a backpack. The refined approach catered for least possible number of variables and parameters in the optimization process and was established based on parameterized muscle gains constraining the lower boundary conditions of trunk muscle coactivations. The validity and reliability of the optimal boundary condition were analyzed by using leave-one-out cross-validation and balanced bootstrap resampling methods. The refined approach provided a good estimator in terms of its unbiasedness, consistency, and efficiency for predicting the peak lumbosacral compression force.

5.1 Background

Walking with a backpack carriage generates critical changes in trunk muscle activation and lumbar joint loading (Li and Chow, 2017). Carrying a load posteriorly deactivates the back muscles and activates the abdominal muscles (Devroy et al., 2007; Li and Chow, 2017). A lack of cocontractions in the extensor and flexor muscles signifies a passive load carriage strategy and potential danger to the human biological structure (Devroy et al., 2007). Therefore, evaluation of the load-bearing strategy of the lumbar spine during walking with a backpack carriage is essential.

Dreischarf et al. (2016) have studied in vivo and computational models for predicting lumbar spine loads and Mohammadi et al. (2015) have applied an EMG-assisted optimization (EMGAO) approach (Amarantini et al., 2010; Cholewicki et al., 1995; Cholewicki and McGill, 1994; Gagnon et al., 2001, 2011) to estimate trunk muscle force and lumbar joint loading. The modeling approach integrates the building blocks of alternative EMG-assisted (EMGA) approaches (Gagnon et al., 2001; Granata and Marras, 1995a; Marras and Granata, 1997; McGill and Norman, 1986) by accounting for biological sensitivity to agonist–antagonist muscle cocontraction activities as well as optimization (OPT) approaches (Bean et al., 1988; Buchanan and Shreeve, 1996; Crowninshield and Brand, 1981; Dul et al., 1984; Hajihosseinali et al., 2014; Stokes and Gardner-Morse, 2001) by achieving the constraints of perfect joint moment equilibrium during dynamic activities. However, collection and processing data numerically is complicated (Cholewicki et al., 1995). The delicate balance between the extents to which conceptual elements and physiological details are excluded or included depends on choosing the possible and easiest model to use that can enable maximal insight into the identified phenomenon required for explanation (Wagner et al., 2012). A refined EMGAO (REMGAO) approach based on an anchored template (a fundamental model with built-in conceptual elements of the identified phenomenon) that caters for the least possible number of variables and parameters (Full and Koditschek, 1999) is worthy of development to enhance the efficacy of data collection and processing.

Trunk muscle force and lumbar joint loading predicted by using EMGAO approaches vary substantially depending on the magnitude of correction factors (muscle gains) introduced in the optimization process (Mohammadi et al., 2015). The boundary conditions of muscle gains restrict the effects of optimization and hinder the flexibility of adjustment for muscle force estimation because a less accurately externally measured resultant joint moment yields a smaller range of boundary conditions to be used to estimate the target joint load (Zheng et al., 1998). The boundary condition settings for the anchored template of the REMGAO approach are vital for estimating muscle force and joint load with degrees of accuracy comparable to those of alternative EMGAO approaches. Essentially, the REMGAO approach was developed to introduce an optimal parametric gain to constrain the boundary conditions of individual muscle forces needed to counterbalance the net moment induced at the lumbar joint.

This part of the study developed a computationally efficient and minimally simple REMGAO approach under the optimal boundary conditions and with the least possible number of variables and parameters for predicting the lumbar joint loading during walking with backpack carriage. The hypothesis of this part of the study was that a predictive model could be identified and adopted to determine the convergence of optimal parametric gains within a range of boundary condition for REMGAO approach along a continuous backpack-loading spectrum.

5.2 Methods

5.2.1 Input data

Anthropometric, kinematic, kinetic, and EMG data collected by the experiment illustrated in Chapter 4 to evaluate the effects of backpack load on trunk muscle activation and lumbosacral joint loading in 10 healthy male adults (refer to Table 4.1 in Chapter 4) during walking were used as input data in this part of the study. Briefly, the participants walked barefoot at their preferred speeds along a walkway embedded with three force platforms carrying no-load (0% BW) and different backpack loads at 5%, 10%, 15%, and 20% BW. Each participant conducted three successful trials (three consecutive clean strikes on the force platforms) for each loading condition. Lower body movements, ground reaction forces, and trunk muscle activity were acquired at sampling rates of 100, 2000, and 2000 Hz, respectively, by using a synchronized system with eight-camera motion analysis (Qualisys, Gothenburg, Sweden), a force platform (Model 4060-10, Bertec Corporation, Columbus, OH, USA), and wireless surface EMG system (Trigno, Delsys Inc., Boston, MA, USA). Raw kinematic, kinetic, and EMG data were filtered. The lumbosacral compression force profile of each identified gait cycle (between two consecutive instances of left heel contacts) during each successful walking trial was estimated using an integrated computation program formulated with a lower body inverse dynamic algorithm (Plamondon et al., 1996), upper body trunk musculoskeletal architecture (Cholewicki et al., 1995; Gagnon et al., 2001; McGill and Norman, 1986), and EMGAO biomechanical model (Cholewicki et al., 1994; Gagnon et al., 2001). Each force profile was time normalized to 101 points. In total, 150 force profiles (10 participants × 5 loading conditions × 3 trials) were obtained for data analysis.

5.2.2 EMG-assisted optimization approach

An EMG-assisted optimization (EMGAO) approach (Table 5.1a) was used as the baseline biomechanical model to estimate the lumbosacral joint compression force. Initial muscle force was predicted based on a nonlinear force–EMG relationship with one predictive variable (EMG) and three parameters, namely EMG at maximal voluntary contraction (EMG_{MVC}), muscle cross-sectional area (A), and maximum muscle intensity (σ_{max}). Passive muscle force was ignored because of its low magnitude in upright stance posture (Cholewicki et al., 1995). The smallest possible muscle gain, g_i $(i = 1,2,\dots,m)$, which adjusts the initial muscle force prediction to perfectly counterbalance the moment generated in the lumbar spine, was figured out using an EMGAO model (Cholewicki and McGill, 1994), with the exception that to ensure adequate prior EMG-based muscle force estimates, the lower g_i bounds were fixed at 0.5 (instead of 0) for all trunk muscles (Gagnon et al., 2001). When $g = 1$, no adjustment of the initial estimated muscle force was required.

5.2.3 *Refined EMG-assisted optimization approach*

A refined EMG-assisted optimization (REMGAO) approach (Table 5.1b) was developed based on the refinements of the aforementioned EMGAO approach. Muscle force was predicted based on one parameter (EMG_{MVC}) and one predictive variable (EMG) by integrating the maximum muscle intensity and cross-sectional muscle area into parameterized muscle gain, k, thereby formulating a least possible parametric linear muscle force and EMG relationship. The cost function was established to obtain the least possible parameterized muscle gain adjustments. The parametric muscle variable, k_i $(i = 1,2, \ldots, m)$, was derived as the product of parametric gain, cross-sectional muscle area, and maximum muscle intensity ($CA_i\sigma_{max}$). When $k_i = A_i\sigma_{max}$, no adjustment was needed. The optimal parametric gain was figured out in a range of settings of C as the optimal boundary condition ($k_i > C_{optimal}A_i\sigma_{max}$) for the REMGAO approach that predicted a comparable lumbar joint force than did the EMGAO approach.

Table 5.1 Formulations of (a) EMGAO and (b) REMGAO approaches. F_i , M_i , A_i , EMG_i , EMG_{MVC_i} , g_i , and k_i represent the predicted force, predicted moment, physiological cross-sectional area, EMG amplitude, EMG_{MVC}, gain (for EMGAO), and parameterized gain (for REMGAO) of muscle i, respectively. $\vec{r}_i$ and $\vec{u}_i$ are the moment arm vector and line of action unit vector of muscle i, respectively. σ_{max} is the maximum muscle intensity. $C_{optimal}$ is the optimal parametric gain of the REMGAO approach. x, y, and z are right-handed orthogonal anatomical axes in posterior–anterior, inferior–superior, and medial–lateral directions. $M_{i.x}$, $M_{i.y}$, and $M_{i.z}$ are internally predicted moments generated by muscle i on the x, y, and z axes, respectively. $M_{e.x}$, $M_{e.y}$, and $M_{e.z}$ are externally measured joint moments on the x, y, and z axes, respectively. n is the number of muscles adopted in the optimization process.

		(a) EMGAO	(b) REMGAO
Predicted muscle force and joint moment		$F_i = g_i A_i \sigma_{max} \left(\frac{EMG_i}{EMG_{MVC_i}}\right)^{\frac{1}{1.3}}$	$F_i = k_i \left(\frac{EMG_i}{EMG_{MVC_i}}\right)$
		$\vec{F}_i = F_i \vec{u}_i$ $\vec{M}_i = \vec{r}_i \times \vec{F}_i$	
Optimization model	Cost function	$\sum_{i=1}^{m} M_i (1 - g_i)^2$	$\sum_{i=1}^{m} M_i (1 - \frac{k_i}{A_i \sigma_{max}})^2$
		where $M_i = \sqrt{{M_{i.x}}^2 + {M_{i.y}}^2 + {M_{i.z}}^2}$	
	Constraint	$\sum_{i=1}^{m} g_i M_{i.x} = M_{e.x}$ $\sum_{i=1}^{m} g_i M_{i.y} = M_{e.y}$ $\sum_{i=1}^{m} g_i M_{i.z} = M_{e.z}$	$\sum_{i=1}^{m} \left(\frac{k_i}{A_i \sigma_{max}}\right) M_{i.x} = M_{e.x}$ $\sum_{i=1}^{m} \left(\frac{k_i}{A_i \sigma_{max}}\right) M_{i.y} = M_{e.y}$ $\sum_{i=1}^{m} \left(\frac{k_i}{A_i \sigma_{max}}\right) M_{i.z} = M_{e.z}$
	Boundary condition	$g_i \geq 0.5$	$k_i \geq C_{optimal} A_i \sigma_{max}$
	Nonnegativity	$F_i \geq 0$	
	Index	$i = 1, 2, \ldots, m$	

5.2.4 Optimal parametric gain

A computational algorithm was developed to identify the optimal parametric gain, $C_{optimal}$, along the continuous loading spectrum between 0% and 20% BW (Figure 5.1). (I) Anthropometric and filtered kinematic, kinetic, and EMG data of each participant $P\ (1, 2, \ldots, n)$ and trial $T\ (1, 2, 3)$ for each loading condition L (0%, 5%, 10%, 15%, and 20% BW) were retrieved from the experimental dataset. The lumbosacral joint compression force profile within an identified gait cycle in each walking trial was predicted and time-normalized in the discrete data point D (1, 2, …, 101) through the REMGAO approach with parametric gain C set at 0.5, 0.6, ... 1.5 (Zheng et al., 1998) for each loading condition. (II) Under each loading condition and parametric gain setting, the root mean square difference (RMSD) between the force profiles predicted by the EMGAO and REMGAO models were computed (equation (1)) and its regression (RRMSD) with respect to the parametric gain setting was identified (equation (2)). (III) The minimum RRMSD for each loading condition was set as the target level. A Chebyshev variant goal programming model (Li and Chow, 2016) was used to figure out the optimal parametric gain along the continuous loading spectrum.

$$\mathrm{RMSD_{C.L}} = \sqrt{\frac{1}{303n}\sum_{P=1}^{n}\sum_{T=1}^{3}\sum_{D=1}^{101}\left(\frac{\mathrm{REMGAO_{L.P.T.D}} - \mathrm{EMGAO_{L.P.T.D}}}{\mathrm{EMGAO_{L.P.T.D}}}\right)^2} \qquad (1)$$

where $C = 0.5, 0.6, \ldots\ , 1.5$ and $L = 0, 5, 10, 15, 20$

$$\mathrm{RRMSD_L} = f_L(C) \qquad \text{where } 0.5 \le C \le 1.5 \text{ and } L = 0, 5, 10, 15, 20. \qquad (2)$$

(I) Compute compression force profile (CFP) of an identified gait cycle

- Retrieve anthropometric as well as filtered kinematic, kinetic and EMG data
  - Participant, P=1, 2, …, n
  - Backpack load, L= 0, 5, 10, 15, 20 % BW
  - Trial, T= 1, 2, 3
- Apply inverse dynamic, musculoskeletal and REMGAO models
  - Parametric gain setting, C = 0.5, 0.6, … , or 1.5
- Time-normalize each CFP into 101 points
- **Output**
  - 15n CFPs for each C

↓

(II) Calculate root mean square difference (RMSD) between the CFPs predicted by REMGAO and EMGAO models

- Pool 3n CFPs for each C under each L
- Calculate RMSD for each C under each L (refer to equation (1) in the text)
- Fit regression to 11 RMSDs for each L (refer to equation (2) in the text)
- **Output**
  - 5 regressions for predicting RMSD by C in the range between 0.5 and 1.5.

↓

(III) Caculate optimal parametric gain for current data set

- Set minimum RMSD for each regression as target levels
- Apply Chebyshev variant goal programming model
- Found C with minimum deviations from the 5 target levels
- **Output**
  - The optimal parametric gain, $C_{optimal}$ for L between 0 and 20% BW

Figure 5.1 Computational algorithm for determining the optimal parametric gain of the REMGAO approach.

5.2.5 Sensitivity analysis

Leave-one-out cross-validation and balanced bootstrap resampling algorithms were adopted to investigate the validity and reliability of the optimal parametric gain determined based on the experimental dataset in this part of the study (Table 5.2). (a) Predictive or adjusted optimal parametric gain was estimated using a leave-one-out cross-validation method (Kim et al., 2012). The 10 datasets were resampled into 10 matched dataset pairs, each consisting of one of the ten datasets selected as a testing set and a training set composed of the other nine. Optimal parametric gains were determined for each testing and training set. The predictive optimal parametric gain was identified by the training set that deviated least from the corresponding testing set. (b) The confidence interval of the optimal parametric gain was estimated using a balanced bootstrap resampling method (Hung et al., 2011). Datasets of 3n (n = number of participants) force profiles for each loading condition were duplicated into 3n samples yielding a total of $9n^2$ samples. Subsequently, 3n samples were randomly chosen from the $9n^2$ samples and the total 3n resampled dataset with 3n force profiles was obtained. For each resampled dataset, the optimal parametric gain was determined. The bootstrap resampling distribution was developed based on the 3n optimal parametric gains. The confidence interval of the optimal parametric gain was determined using this distribution.

Table 5.2 Validity and reliability of the optimal parameter gain. (a) Leave-one-out cross-validation ; (b) balanced bootstrap resampling algorithms for determining the predictive and confidence intervals of the optimal parametric gain, respectively.

<table>
<tr><th></th><th colspan="2">(a) Leave-one-out cross-validation algorithm</th><th colspan="4">(b) Balanced bootstrap resampling algorithm</th></tr>
<tr><td>Input</td><td colspan="2">n participants (P_i) → n data sets
$i = 1, 2, \dots, n$</td><td colspan="4">n participants → 3n compression force profiles (CPFs)
for each backpack load condition: L=0, 5,10, 15, 20</td></tr>
<tr><td rowspan="2">Step 1</td><td colspan="2">Resample data into n pairs of testing and training sets</td><td colspan="4" rowspan="2">Duplicate 3n sets of 3n CPFs into $9n^2$ CPFs for each loading condition</td></tr>
<tr><td>Testing set: P_1
Testing set: P_2
.
.
Testing set: P_n</td><td>Training set: all P_i excluding P_1
Training set: all P_i excluding P_2
.
.
Training set: all P_i excluding P_n</td></tr>
<tr><td rowspan="2">Step 2</td><td colspan="2">Compute optimal parametric gains for all testing and training sets by the computation algorithm illustrated in Figure 1</td><td colspan="4">Resample 3n data sets each containing randomly selected 3n CPFs from the $9n^2$ CPFs created in Step 1 for each backpack load condition</td></tr>
<tr><td>$C_{optimal.te.i}$
testing set i, for $i = 1, 2, \dots, n$</td><td>$C_{optimal.tr.i}$
training set i, for $i = 1, 2, \dots, n$</td><td>Set 1</td><td>Set 2</td><td>..........</td><td>Set 3n</td></tr>
<tr><td rowspan="2">Step 3</td><td colspan="2" rowspan="2">Calculate the absolute differences between optimal parametric gains of matched pairs of testing and training sets
$\text{abs}\left[\frac{(C_{optimal.tr.i} - C_{optimal.te.i})}{C_{optimal.te.i}}\right]$, for $i = 1, 2, \dots, n$</td><td colspan="4">Compute optimal parametric gains for all resampled data sets by the computation algorithm illustrated in Figure 1</td></tr>
<tr><td>$C_{optimal.1}$</td><td>$C_{optimal.2}$</td><td>.........</td><td>$C_{optimal.3n}$</td></tr>
<tr><td>Step 4</td><td colspan="2">Determine the matched pair with minimum absolute difference</td><td colspan="4">Build the distribution of optimal parametric gain based on $C_{optimal.i}$ for $i = 1, 2, \dots, 3n$</td></tr>
<tr><td>Step 5</td><td colspan="2">Set $C_{predictive.optimal} = C_{optimal.tr.i}$
where i is the index of the matched pair determined in step 4</td><td colspan="4">Find the confidence interval of optimal parametric gain ($C_{optimal}$) based on the balanced bootstrap resampling distribution created in Step 4</td></tr>
<tr><td>Output</td><td colspan="2">$C_{predictive.optimal}$</td><td colspan="4">$C_{confidence\_interval.optimal}$</td></tr>
</table>

5.2.6 Data reduction and statistical analysis

The number of muscles (m) and participants (n) and were set as 12 and 10, respectively. All data filtering, computation, and optimization processes were implemented using MATLAB 2016b software (MathWorks, Inc., Natick, MA, USA). Chebyshev variant goal programming problems were solved using LINGO 10.0 software (Lindo Systems Inc., Chicago, IL, USA). Statistical analyses and the comparability of the REMGAO approach under the optimal boundary condition for predicting lumbosacral compression force while walking with a backpack carriage were compared using the EMGAO approach (Cholewicki and McGill, 1994; Gagnon et al., 2001) and two fundamental approaches: double linear optimization (DLOPT; Bean et al., 1988; Gagnon et al., 2001) and single linear optimization (SLOPT; Bean et al., 1988) (Table 5.3). SLOPT is a fundamental template with no boundary constraints for individual muscle intensity, DLOPT is a template anchored in physiological constraint of a least possible intensity boundary for all muscles. Two complete sets of compression force profiles and RMSDs were also computed by the two approaches. Mean deviations of peak compression forces predicted by the REMGAO, DLOPT and SLOPT from EMGAO approaches were figured out. Repeated measure ANOVA with backpack load as within-subject factor was adopted to analyze the effect of backpack load on the mean deviations, one-sample t-test was adopted to analyze the deviations from zero, Shapiro-Wilk test was used to justify the assumption of normality for the data sets of deviations and optimal parametric gains of the bootstrap resamples (SPSS version 24.0, IBM Inc., Chicago, IL, USA). Statistical significance was set at $p<0.05$.

Table 5.3 Formulation of (a) DLOPT and (b) SLOPT approaches. F_i , M_i , and A_i represent the force, moment, and physiological cross-sectional area of muscle i, respectively. $\vec{r}_i$ and $\vec{u}_i$ are the moment arm vector and line of action unit vector of muscle i, respectively. σ_{max} is the maximum muscle intensity. G is the subject-invariant gain in the first DLOPT process and G* is the optimized gain setting for the upper bound of the muscle force in the second process. x, y, and z are right-handed orthogonal anatomical axes in posterior–anterior, inferior–superior, and medial–lateral directions. $M_{i.x}$, $M_{i.y}$, and $M_{i.z}$ are internally predicted moments generated by muscle i on the x, y, and z axes, respectively. $M_{e.x}$, $M_{e.y}$, and $M_{e.z}$ are externally measured joint moments on the x, y, and z axes, respectively. n is the number of muscles adopted in the optimization process.

<table>
<tr><th colspan="2"></th><th>(a) DLOPT</th><th>(b) SLOPT</th></tr>
<tr><td rowspan="4">First optimization process</td><td>Cost function</td><td>G</td><td>$\sum_{i=1}^{m} F_i$</td></tr>
<tr><td>Constraint</td><td colspan="2">$\vec{F}_i = F_i\vec{u}_i \quad , \quad \vec{M}_i = \vec{r}_i \times \vec{F}_i$
$\sum_{i=1}^{m} M_{i.x} = M_{e.x} \ , \sum_{i=1}^{m} M_{i.y} = M_{e.y}, \sum_{i=1}^{m} M_{i.z} = M_{e.z}$</td></tr>
<tr><td>Boundary condition</td><td>$G \geq \frac{F_i}{A_i\sigma_{max}}$, for $i = 1.2, \ldots, m$</td><td>-</td></tr>
<tr><td>Nonnegativity</td><td colspan="2">$F_i \geq 0$, for $i = 1.2, \ldots, m$</td></tr>
<tr><td rowspan="4">Second optimization process</td><td>Cost function</td><td>$\sum_{i=1}^{m} F_i$</td><td rowspan="4">-</td></tr>
<tr><td>Constraint</td><td>$\vec{F}_i = F_i\vec{u}_i \quad , \quad \vec{M}_i = \vec{r}_i \times \vec{F}_i$
$\sum_{i=1}^{m} M_{i.x} = M_{e.x} \ , \sum_{i=1}^{m} M_{i.y} = M_{e.y}, \sum_{i=1}^{m} M_{i.z} = M_{e.z}$</td></tr>
<tr><td>Boundary condition</td><td>$F_i \geq G^* A_i \sigma_{max}$, for $i = 1, 2, \ldots, m$
(G^* is the solution of the first optimization)</td></tr>
<tr><td>Nonnegativity</td><td>$F_i \geq 0$, for $i = 1.2, \ldots, m$</td></tr>
</table>

5.3 Results

5.3.1 Root mean square difference

The RMSD for each backpack load for the REMGAO, DLOPT, and SLOPT approaches and each parametric gain setting for the REMGAO approach were determined based on the experimental dataset (Table 5.4a). The mean RMSDs were within the ranges 5.8%–13.7%, 15.4%–17.7%, and 24.5%–27.1% for the REMGAO, DLOPT, and SLOPT approaches, respectively.

The RMSD regressions for each backpack were determined based on a second order polynomial model (Table 5.4b). The R^2 values for the five RMSD regressions were all greater than 0.98. The RMSDs converged to minima between 6.0 and 7.1 at parametric gain settings between 0.93 and 1.03.

5.3.2 Optimal parametric gain

The Chebyshev variant goal programming model was developed based on the five RMSD regressions and the target levels for minimum RMSD of 7.1, 6.0, 6.2, 6.4, and 6.3 for the backpack loads of 0%, 5%, 10%, 15%, and 20% BW, respectively (Table 5.4c). The optimal boundary condition $(\mathrm{C_{optimal}})$ was determined as 0.980.

Table 5.4 Evaluation of the RMSDs of the REMGAO, DLOPT, and SLOPT approaches for each backpack load condition. (a) Comparison of RMSDs with those of the EMGAO approach; (b) RMSD regressions for the REMGAO approach; (c) Chebyshev variant goal programming model for determining the $C_{optimal}$ of the REMGAO approach.

(a) RMSDs comparing with EMGAO approach

	C	RMSD* x 100%				
		0% BW	5% BW	10% BW	15% BW	20% BW
REMAGO	0.5	11.1	11.4	10.7	10.7	10.5
	0.6	9.8	9.9	9.3	9.2	9.2
	0.7	8.7	8.5	8.1	7.8	8.1
	0.8	7.8	7.3	7.0	6.8	7.1
	0.9	7.2	6.3	6.3	6.2	6.4
	1.0	7.0	5.8^	6.0	6.3	6.1
	1.1	7.2	5.9	6.3	7.0	6.3
	1.2	7.9	6.5	7.2	8.2	7.0
	1.3	8.9	7.6	8.4	9.8	8.1
	1.4	10.2	9.0	9.8	11.7	9.4
	1.5	11.7	10.5	11.4	13.7#	10.9
DLOPT	-	17.7##	16.6	16.5	17.1	15.4^^
SLOPT	-	27.1###	26.0	25.0	25.7	24.5^^^

* mean of 3 walking trials for each C under each backpack loading condition
^ minimum and # maximum RMSDs of REMGAO approach
^^ minimum and ## maximum RMSDs of DLOPT approach
^^^ minimum and ### maximum RMSDs of SLOPT approach

(b) Regressions of RMSD for REMGAO approach

Load (% WB)	Second order polynomial regression	Minimum RMSD (target of goal programming model)
0	$17.452C^2 - 34.389C + 24.051$ (R^2=0.996)	7.1 at C=0.985
5	$20.309C^2 - 41.785C + 27.520$ (R^2=0.989)	6..0 at C=1.029
10	$20.078C^2 - 39.523C + 25.670$ (R^2=0.991)	6.2 at C=0.984
15	$23.335C^2 - 43.524C + 26.706$ (R^2=0.994)	6.4 at C=0.933
20	$18.312C^2 - 36.363C + 24.324$(R^2=0.992)	6.3 at C=0.993

(c) Chebyshev variant goal programming model for determining $C_{optimal}$ of REMAGO approach

Cost Function	H
Constraints	$17.452C^2 - 34.389C - D_{00}^{+} + D_{00}^{-} = -16.941$ $20.309C^2 - 41.785C - D_{05}^{+} + D_{05}^{-} = -21.493$ $20.078C^2 - 39.523C - D_{10}^{+} + D_{10}^{-} = -19.450$ $23.335C^2 - 43.524C - D_{15}^{+} + D_{15}^{-} = -20.295$ $18.312C^2 - 36.363C - D_{20}^{+} + D_{20}^{-} = -18.052$ $0.141 * D_{00}^{+} \leq H$ $0.166 * D_{05}^{+} \leq H$ $0.161 * D_{10}^{+} \leq H$ $0.156 * D_{15}^{+} \leq H$ $0.159 * D_{20}^{+} \leq H$
Boundaries	$0.5 \leq C \leq 1.5$
Non-negativity	$D_{00}^{-}, D_{00}^{+}, D_{05}^{-}, D_{05}^{+}, D_{10}^{-}, D_{10}^{+}, D_{15}^{-}, D_{15}^{+}, D_{20}^{-}, D_{20}^{+} \geq 0$
Solution	$C_{optimal} = 0.9798$
Abbreviations	D_i^{+}: positive deviation from target level of backpack loading condition i D_i^{-}: negative deviation from target level of backpack loading condition i i = 00 (0% BW), 05 (5% BW), 10 (10% BW), 15 (15% BW), 20 (20% BW)

The predictive optimal parametric gain ($C_{predictive.optimal}$) was determined as 0.979 based on the leave-one-out cross-validation algorithm (Table 5.5). The optimal parametric gains of the 10 predicting training sets ranged between 0.974 and 1.019. The optimal predicting training set value of 0.979 underestimated the corresponding predicted testing set value of 0.995 by 1.6%.

Table 5.5 Determination of predictive $C_{optimal}$ by leave-one-out cross-validation algorithm

Test set	**$C_{optimal}$**	**Training set**	**$C_{optimal}$**	**Deviation (%)**
Participant		All participants		-10.7
1	1.086	except 1	0.979	-10.7
2	0.796	except 2	0.983	18.7
3	0.875	except 3	0.990	11.5
4	1.220	except 4	0.974	-24.6
5	1.114	except 5	0.982	-13.1
6	0.995	except 6	0.979*	-1.6^
7	0.780	except 7	0.979	19.8
8	0.876	except 8	1.019	14.3
9	0.929	except 9	0.998	6.9
10	0.832	except 10	0.981	14.9

* Predictive optimal parametric gain
^ Least deviations between test and training sets

The 95% confidence interval of the optimal parametric gain ($C_{confidence\_interval.optimal}$) was determined as (0.965, 0.985) based on the balanced bootstrap resampling algorithm (Table 5.6). The optimal parametric gains of the 30 bootstrap resampled datasets ranged between 0.937 and 1.035 with a mean of 0.975 and standard deviations of 0.0262. The optimal parametric gain resampled data normality was validated using the Shapiro–Wilk test (p = 0.981).

Table 5.6 Determination of confidence interval of $C_{optimal}$ by balanced bootstrap resampling algorithm

Resampling set	$C_{optimal}$
1, 2 ,3, 4, 5, 6	0.940, 0.981, 1.008, 0.989, 0.991, 0.990
7, 8, 9, 10, 11, 12	0.945, 0.992, 0.954, 1.035, 0.954, 0.971
13, 14, 15, 16, 17, 18	0.975, 1.007, 0.978, 0.986, 0.962, 0.942
19, 20, 21, 22, 23, 24	1.018, 0.972, 0.970, 1.000, 0.954, 0.975
25, 26. 27, 28, 29, 30	0.925, 1.035, 0.971, 0.957, 0.967, 0.937
Overall mean = 0.975, SD = 0.0262, 95% confidence interval (0.965, 0.985) Shapiro-Wilk normality test: p = 0.981	

5.3.3 Statistical evidence

The peak compression force deviations of REMGAO, DLOPT, and SLOPT approaches from EMGAO approach were figured out for each force profile (Appendix 5.1). No significant backpack load effects on the mean peak compression force deviations (from EMGAO approaches) were observed for the REMGAO, DLOPT, or SLOPT approaches ($p = 0.486$, 0.618, and 0.371, respectively). On average, no significant difference (mean = −0.1%; standard error of the mean (SEM) = 0.3%; p = 0.862) was observed between the predictions of peak compression force calculated using the EMGAO and REMGAO approaches; however, the DLOPT and SLOPT predictions significantly underestimated ($p < 0.001$) the peak compression force by an averages of 5.1% (SEM = 0.7%) and 19.2% (SEM = 0.5%), respectively (Table 5.7a). The normality assumptions of peak compression force deviations for the three approaches were validated using the Shapiro–Wilk test ($p = 0.275$, 0.320, and 0.471, for the REMGAO, DLOPT, and SLOPT approaches, respectively).

The overall peak compression force RMSDs of REMGAO, DLOPT, and SLOPT were determined as 4.2%, 9.9%, and 20.3%, respectively, and the overall compression force profile RMSDs were determined as 6.2%, 16.3%, and 25.6%, respectively (Table 5.7b). The REMGAO approach predicted a peak compression force and compression force profile with the lowest RMSDs compared with the EMGAO approach.

Table 5.7 Analyses of deviations in predicted lumbosacral joint compression forces compared with those of the EMGAO approach. (a) Mean deviations of peak compression forces; (b) overall RMSDs of peak compression force and compression force profile.

(a) Deviations (%) of predicted peak compression forces

Mean of 30 trials (10 participants x 3 trials)	Backpack load					Pooled mean (standard error of the mean) of all participants (150 trials)	Statistical tests		
							Repeated measures ANOVA (effect of backpack load on mean deviation)	One-sample t-test	
	0% BW	5% BW	10% BW	15% BW	20% BW			Pooled mean deviation = 0	Shapiro-Wilk normality test
REMGAO	0.7	0.2	0.4	-0.4	-1.2	-0.1 (0.3) ##	p=0.486#	p=0.862##	p=0.275###
DLOPT	-4.6	-7.1	-5.8	-3.7	-4.2	-5.1 (0.7) *	p=0.618#	p<0.001*	p=0.320###
SLOPT	-17.8	-18.6	-21.5	-18.8	-19.5	-19.2 (0.5) *	P=0.371#	p<0.001*	p=0.471###

The effects of backpack load on the mean deviations were not significant for all three approaches
On average, there was no significant difference in predicted peak lumbosacral compression forces between REMGAO and EMGAO approaches
* On average, DLOPT and SLOPT approaches significantly underestimated peak lumbosacral compression force by 5.1% and 19.2% respectively
Normality of data was assumed

(b) Overall RMSDs (%) of predicted peak compression forces and compression force profiles

	Peak compression force	Compression force profile
REMGAO	4.2^	6.2^^
DLOPT	9.9	16.3
SLOPT	20.3	25.6

^ REMGAO approach predicted the peak compression force with least RMSD from EMGAO
^^ REMGAO approach predicted the compression force profile with least RMSD from EMGAO

5.4 Discussion

5.4.1 Computation framework

In this part of the study, a REMGAO approach was developed with the least possible number of variables and parameters for predicting the lumbar joint loading while walking with a backpack with a loaded backpack up to a maximum of 20% BW. The novelty of the REMGAO approach lay in the introduction of an optimal parametric gain constraining the boundary condition of the individual muscle forces demanded to counterbalance the net moment induced at the lumbar joint. Compared with the EMGAO approaches developed in previous studies, the REMGAO approach is a numerically efficient and simple computation framework.

The complexity of the EMGAO approaches originated from the formulation of the EMG–force relationship. The most sophisticated EMGAO approach caters for an EMG–force relationship with the following four time-dependent variables: correction gain factor, EMG amplitude, and force length- and force velocity-modulated factors. In addition, the following three time-invariant parameters were also included: EMG amplitude at MVC, physiological muscle cross-sectional area, and maximum muscle intensity (Gagnon et al., 2001; Mohammadi et al., 2015). Alternatively, EMGAO approach could adopt a subject-invariant five-parameter formulation with the following five muscle parameters: the minimizer of moment prediction errors, minimizer of the moment phase shift, stresses for flexors, stresses for extensors, and linear relationship of force-normalized EMG (Nussbaum and Chaffin, 1998). The complexity of considering surplus variables and parameters might not improve the prediction accuracy; for example, the force velocity factor showed no significant predictability under moderate speeds (Nussbaum et al., 1999) and only slight corrections were required (Granata and Marras, 1995b; McGill and Norman, 1986). The REMGAO approach simplifies the EMGAO approach by allowing two variables (modulated parametric gain factor and EMG amplitude) and one parameter (EMG amplitude at MVC) in the optimization process by constraining the boundary conditions with the optimal parametric gains.

5.4.2 EMG–force relationship

The REMGAO approach applies a linear EMG–force relationship to facilitate simplify formulation of the computation process. The EMG–force relationship is the fundamental physiological building block of the EMG-based biomechanical model for predicting lumbar joint loading and trunk muscle force (Marras and Granata, 1997). The observation of a nonlinear or linear EMG–force relationship depends on the different recruitment maneuvers of the abdominal and back muscles during trunk extension and flexion activities (Anders et al., 2008). In contrast to the nonlinear EMG–force relationship-based EMGAO approach (Cholewicki et al., 1995) and the linear EMG–force relationship-based EMGA approach (McGill and Norman, 1986), which predicted similar compression forces at the lumbar joint, nonlinear EMG–force relationship-based EMGAO approach and the linear EMG–force relationship-based REMGAO approach in this part of the study predicted no significant differences in the mean peak lumbosacral compression force. The linear

EMG–force relationship adopted by the REMGAO approach was thereby justified.

5.4.3 Unbiased estimation

The unbiasedness, consistency, and efficiency of the mean peak lumbosacral compression force predicted by the REMGAO approach met the standard requirements of a good estimator (Steiger, 2000) of the mean estimated using the EMGAO approach. The standard error of the mean deviations for the REMGAO, DLOPT, and SLOPT approaches were quite small (lower than 1%), but only the mean predicted by the REMGAO approach was not significantly different from that predicted by the EMGAO approach. Moreover, the deviation of the REMGAO approach was assumed to be normally distributed with a mean of approximately 0. The mean predicted by the REMGAO approach was unbiased, consistent, and efficient, and thus serves as a good estimator for the mean predicted by the EMGAO approach. Cholewicki et al. (1995) validated that the EMGAO approach predicted insignificant and minimal differences in the lumbar joint (L4–L5) compression force compared with the EMGA approach (McGill and Norman, 1986). In addition, the mean peak lumbosacral compression force predicted by the REMGAO approach may be a good estimator of the peak force predicted by the EMGA approach.

5.4.4 Validity and reliability

The setting of the parametric muscle gain between 0.5 and 1.5 was based on a previous study on developing biomechanical model in knee joint loads (Zheng et al., 1998) and validated by the convergence of optimal parametric gain by this part of the study.

The 95% confidence interval (0.965, 0.985) of the bootstrap resampling distribution covered both the predictive optimal parametric gain (0.979) and the optimal parametric gain (0.980) determined by the leave-one-out algorithm and from the experimental data and, respectively. The confidence interval of the optimal boundary condition developed using the balanced bootstrap resampling distribution is a reliable range covering the optimal boundary condition setting; however, it should not be used to validate the condition for individual participants, but rather to develop sensitivity and predictive analyses (Lenhoff et al., 1999; Mikshowsky et al., 2017) for overall estimation in future experimental studies estimating lumbosacral compression force profiles while walking with a loaded backpack up to a maximum of 20% BW. The predictive optimal parametric gain of 0.979 ensured that the lower force boundary allocated to individual muscles was larger than 0.979 times the product of cross-sectional muscle area, maximum muscle intensity, and normalized EMG. A previous study using the EMGA approach reported that the average muscle gain was between 0.940 and 1.354 during dynamic and static extension and flexion exercises (Marras and Granata, 1997). The optimal boundary condition of muscle force in the REMGAO approach is an appropriate lower boundary condition for walking with a backpack carriage.

The REMGAO approach is an anchored template that predicts an unbiased peak lumbosacral compression force for walking with a backpack carriage. The effect of neglecting antagonistic cocontraction on the lumbar spine load was minor at the lumbosacral level (Van Dieen and Kingma, 2005) and much lower compression forces (23%–43%) were detected at levels L4–L5 (Cholewicki et al., 1995) compared with the optimization- and EMG-based approaches. The prediction errors depended on the nature of the tasks conducted and constraints introduced in the optimization process. The DLOPT and SLOPT approaches in this study underestimated the peak compression forces by 5.1% and 19.2%, respectively, possibly indicating the error range predicted by the optimization-based approach for walking with a backpack carriage.

5.4.5 Limitations

The predictive optimal parametric gain of 0.979 was based on the refinement of an EMGAO approach and a computation algorithm specifically for walking under various load conditions between no load and a loaded backpack up to 20% BW. In addition, the comparability of the REMGAO approach was justified by the prediction of compression force in the lumbosacral joint. The generalization of the REMGAO approach to other human locomotion and joints in the upper spine level and lower limb is needed for further studies.

5.5 Summary

The REMGAO approach was developed based on the converged optimal parametric gain and is simple in nature, efficient in computation, and unbiased in predicting the lumbosacral compression force of walking with a loaded backpack between 0% and 20% BW. The generalization of the REMGAO approach could be used in biomechanical modeling to estimate the mechanical loads of various human joints during dynamic motions.

CHAPTER 6*

Effects of asymmetric loading on lateral spinal curvature in young adults with scoliosis: A preliminary study

* This chapter has been accepted for publication by the author of this thesis.
Li, S. S. W., & Chow, D. H. K. (2018). Effects of asymmetric loading on lateral spinal curvature in young adults with scoliosis: A preliminary study. Prosthetics and Orthotics International. (In press).

Usual guideline recommends symmetric over asymmetric load carriage. Whether this recommendation is valid for subjects with asymmetric body alignment, such as those with scoliosis, remains unclear. The nature of scoliosis is subject-dependent and time-variant. Interventions are generally employed to prevent the abnormal spinal curvature from progressing. This part of the study aimed at investigating the effects of an asymmetric load carriage on lateral spinal deformity in subject with scoliosis. Photogrammetry was adopted to measure the scoliotic curvature changes in thoracic and lumbar regions in upright stance posture with no-load and a single-strap cross-chest bag loaded up to 12.5% of body weight. A multi-objective goal programming approach was developed for determining the optimal configuration of asymmetric load carriage for patients with scoliosis.

6.1 Background

Scoliosis is a 3-dimensional deformity of the spine (Weinstein et al., 2008). It is classified as idiopathic, neuromuscular, or congenital, and attains its progressive growth spurt during adolescence (Hresko, 2013). Around 2%–4% of adolescents are suffered from idiopathic scoliosis (Carrier et al., 2004; Nie et al., 2008; Wang et al., 2014). A screening reported that the prevalence of adolescent idiopathic scoliosis (AIS) was 2.5% in population group aged less than 19 (Luk et al., 2010). Various long-term impacts of spinal deformity have been validated, including scoliotic curve progression during the adolescent growth spurt (Shi et al.,2011), psychological conditions (Mayo et al., 1994; Sahlstrand et al., 1978; Weinstein et al., 2003), motor control impairment (Byl et al., 1997; Mayo et al., 1994; Nault et at., 2002; Sahlstrand et al., 1978), higher prevalence of back pain (Weinstein et al., 2003), and poorer pulmonary function (Boyer et al., 1996; Muirhead and Conner, 1985). The complex mechanism and unknown aetiology of scoliotic spine growth render preventive treatments and prognostic difficult (Stokes, 2007).

The nature of AIS is subject-dependent and time-variant. Intervention management is commonly applied to prevent the abnormal spinal curvature from progression. Severe AIS (Cobb angle $> 45^0$) demand surgical operations, moderate AIS ($25^0 <$ Cobb angle $< 45^0$) are managed by bracing together with physiotherapy, and mild AIS (Cobb angle $< 25^0$) are prescribed with physiotherapeutic exercises or observation only (Hresko, 2013; Weinstein et al., 2008).

In order to prevent the atypical spinal curvature from progressing, asymmetric exercises have been suggested for patients with scoliosis. Durmala et al. (2003) evaluated the impacts of asymmetric mobilisation on the angle of axial rotation and Cobb angle in 136 patients. After a year of training, results showed a 16%–23% improvement in axial rotation and a 31%–39% reduction in the Cobb angle. Mooney and Brigham (2003) and Mooney et al. (2000) investigated torso rotational strength revealing that the muscle strength of the convex side was stronger than that of the concave. According to the strength test findings, they prescribe the weaker muscles for asymmetric rotational exercise, which reduced the muscle imbalance of all patients with mild scoliosis after 4 months. Except in one case with increased curvature, all spinal curves were remained either unchanged or alleviated by an average of 20%. Numerous studies have evaluated the effectiveness of asymmetric exercises and side-shift in alleviating the scoliotic curve progression rate (Fusco et al., 2011; Mamyama et al., 2001; Maruyama et al., 2003; Mooney and Brigham, 2003; Mooney et al., 2000; Negrini et al., 2012; Negrini and Negrini, 2007; Pascoe et al., 1997) and have confirmed the efficacy of bracing in improving balance and mobility (Fusco et al., 2011). Despite of this positive evidence, compliance with the prescribed training protocol is vital to achieve effective outcomes. However, it is impractical to expect patients with scoliosis to comply with self-administered exercises at home or attend therapy sessions 2–3 times per week.

Common guideline suggests symmetric over asymmetric load carriage. Whether this suggestion is valid for subjects with asymmetric body alignment, such as patients with scoliosis, remains unclear. When patients with scoliosis carry regular symmetric load, such as a backpack, “asymmetric” stresses are generated on their intervertebral endplates, which may induce further asymmetric spine growth and create a vicious cycle of scoliotic spinal progression (Fok et al., 2010). As spine growth was related to applied stress (Den Boer, et al., 1999; Motmans et al., 2006), the prescription of properly controlled asymmetric loads at either side of the body is proposed in this part of the study for postural rectification in patients with scoliosis. The aim was to evaluate the loading configuration (weight and position) of asymmetric load carriage under the hypothesis that properly controlled asymmetric load carriage alleviates the Cobb angle of the affected region of the scoliotic spine. The target corrective measures of both unaffected and affected regions under asymmetric load carriage were treated as a multi-objective problem. The biomechanical effects of asymmetric load carriage of various weights on spinal curvature were evaluated experimentally in AIS patients. This part of the study hypothesised that an appropriate multi-objective goal programming approach could be identified and used to determining the optimal curvature changes in both the unaffected and affected regions of the scoliotic spine under properly controlled asymmetric load carriage up to 12.5% of body weight (BW).

6.2 Methods

6.2.1 Participants

Seven young adults with mild scoliosis were recruited for this part of the study (Table 6.1). All patients with mild scoliosis had a minor lumbar curve and a major thoracic curve (S-curve). Three of them had a right major curve and the others had a left major curve; the mean Cobb angle was 17.4° (SD=3.6°). Patients with scoliosis, who had received bracing, surgical, or any other clinical treatments for scoliosis, as well as those with a history of other musculoskeletal illnesses, were excluded from this part of the study. The Human Research Ethics Committee granted ethical approval and all participants provided written informed consent prior to experimentation.

Table 6.1 Demographics and Cobb angle measurements of the 7 participants.

Participant	Gender	Apex location of major scoliotic S-curve	Upper end vertebra	Lower end vertebra	Cobb angle (degree)	Age (year)	Body weight (kg)
1	F	Right	T4	T10	22.2	21	42.5
2	F	Right	T5	T12	21.9	21	46.3
3	F	Right	T2	T9	17.2	22	50.0
4	M	Left	T3	T11	14.3	20	56.1
5	M	Left	T2	T10	15.9	22	55.2
6	F	Left	T4	T11	12.6	19	47.1
7	F	Left	T4	T10	17.4	16	42.3
Mean (SD)					**17.4**	**20.1**	**48.5**
SD					**3.6**	**2.1**	**5.6**

6.2.2 Procedures

The participants were instructed to maintain a relaxed, erect, and barefoot standing posture with the arms hanging freely at both sides, a gaze fixed on a target placed 2 m ahead at eye level, and the feet at a 30° angle between their long axes with the heels 14 cm apart (Chow et al., 2005a, 2007; Sahlstrand et al., 1978). The participants were asked to carry a single-strap cross-chest shoulder bag weighing 2.5%, 5%, 7.5%, 10%, and 12.5% of their BW at the contralateral and ipsilateral sides relative to the apex of their major scoliotic curve (Chow et al., 2006b). The centre of gravity of the bag was positioned at the anterior superior iliac spine level and on the coronal plane. The spinous processes of the participants were palpated between C7 and L5, and identified externally along the whole spine using circular markers (Aroeira et al., 2011). Digital anterior-posterior photos were taken to record the respective positions of the markers on the spinous processes for each experimental trial (Figure 6.1).

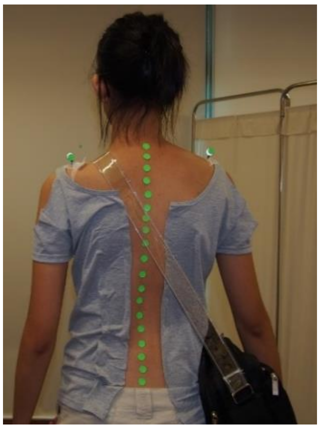


Figure 6.1 Digital anterior-posterior photo of participant 7 with the single-strap cross-chest bag placed on the contralateral side relative to the apex location and loaded at 5% of body weight. Circular markers were affixed to the spinous processes along the spine between C7 and L5.

Although radiographic images and ultrasound method would have demonstrated higher and comparable accuracy, respectively in evaluating the spinal curvatures, digital photos were adopted to avoid radiation and blocking by the strap of the bag across the back. The participants rested for 2–5 minutes between consecutive experimental trials.

For participants 1-6, one trial for the no-load and each loaded condition were conducted at the contralateral and ipsilateral sides relative to the apex. The sequence of the 11 static trials was randomized. The data recorded were used for evaluating the effective side of asymmetric loading for mild scoliosis patients, which features optimal reduction in the major curve.

For participant 7, three trials for the no-load and each loaded condition were conducted on the effective side determined by the results of participants 1-6. The sequence of the 18 static capturing trials was randomized. The data recorded were used for evaluating the optimal asymmetric loading condition for patients with AIS, featuring minimal negative effects on the minor lumbar curve and optimal reduction in the major thoracic curve.

6.2.3 Data analysis

Image processing techniques was implemented to analyse the digital photos taken for all 7 participants. The centroids of 18 circular markers (from C7 to L5) were determined as the spinous processes. A continuous spline curve was then fitted to the 18 points. The positions of the apexes and the respective lower and upper scoliotic end plates were identified. Then, the minor lumbar Cobb angle and the major thoracic Cobb angle were determined.

The major scoliotic curvatures determined for participants 1-6 under different asymmetric loading conditions were evaluated by using repeated measure ANOVA, with asymmetric load on the contralateral and ipsilateral sides as a within-subjects factor, was used to evaluate the means of the major thoracic Cobb angles. The minor lumbar Cobb angle and major thoracic Cobb angle determined for participant 7 under various asymmetric loading conditions were evaluated using One-way ANOVA to test the effect of asymmetric load on the minor and major Cobb angles. Statistical power and level of significance were set at 0.8 and $p=0.05$, respectively. Post hoc pairwise comparisons were based on least squares significant difference (LSD) criterion.

Different trend lines of the mean spinal curvatures were fitted to the data with respect to the loading conditions. Regression models of the lumbar Cobb angle (LCA) and thoracic Cobb angle (TCA) under loading conditions (L, % of BW) were obtained as below:

$$TCA = f_{TCA}(L) \text{ and } LCA = f_{LCA}(L), \text{ where } L = 2.5, 5, 7.5, 10, 12.5$$

6.2.4 Lexicographic goal programming approach

In the context of programming, a goal is defined as the criterion, target, or desired level. A decision analysis in more than one goal is classified as a multi-criterion decision analysis problem. Classical optimization approach solves problems by finding an optimal solution for a unique objective. The lexicographic goal programming (LGP) approach (Ghandforoush, 1993; Li and Chow, 2016) was adopted to determine an optimal solution for various prioritized target levels. Lexicographic goal programming formulates the desire corrective measures of both minor and major curves according to the spinal curvatures of the individual curves. The spinal curvatures were expressed and prioritized as general linear regressions of the configuration parameters of asymmetric physiotherapy exercises. In this part of the study, the lexicographic goal programming model was applied to achieve the prioritized goals individually. The solution of the lexicographic goal programming approach was the optimal configuration parameter of the asymmetric physiotherapy exercise with minimal effects on the minor curve and maximal correction at the major curve of the scoliotic spine. The output of the optimal load was determined through the following sequential optimization processes: Minimize $(D^{+}_{TCA})^{1st}, (D^{+}_{LCA})^{2nd}$

subject to

$$f_{TCA}(L) + D^{-}_{TCA} - D^{+}_{TCA} = TTCA$$

$$f_{LCA}(L) + D^{-}_{LCA} - D^{+}_{LCA} = TLCA$$

$$2.5 \leq L \leq 12.5$$

$$D^{-}_{TCA}, D^{+}_{TCA}, D^{-}_{LCA}, D^{+}_{LCA} \geq 0$$

$$D^{+}_{TCA} = (D^{+}_{TCA})^{*}$$

[for the second LGP only, $(D^{+}_{TCA})^{*}$ is the solution of the first LGP], where

D^{-}_{TCA} represented the negative deviational variable for the TCA,

D^{+}_{TCA} represented the positive deviational variable for the TCA,

D^{-}_{LCA} represented the negative deviational variable for the LCA,

D^{+}_{LCA} represented the positive deviational variable for the LCA,

TTCA represented the target (mean of loaded conditions) TCA, and

TLCA represented the target (mean of loaded conditions) LCA.

TCA = Thoracic Cobb Angle, LCA = Lumbar Cobb Angle

Statistical tests, image processing, and LGP problems were implemented using SPSS 21.0 (IBM Inc., Chicago, IL, USA), MATLAB 2013b (The MathWorks Inc., Natick, Massachusetts, USA), and LINGO 10.0 (Lindo System Inc., Chicago, IL, USA) software, respectively.

6.3 Results

The spinal curvatures were measured using photogrammetry with no load and the bag loaded on the contralateral and ipsilateral sides relative to apex location for participants 1-6 and effective side (contralateral side relative to the apex location) for participant 7 (Appendix 6.1).

6.3.1 Loading on either side of the apex location

The spinal curvatures of participants 1-6 were evaluated using the repeated measure ANOVA (Table 6.2). When the load was placed on ipsilateral side relative to the apex location, there were no significant changes (p=0.773) in the major thoracic Cobb angles. However, placing the load on contralateral side relative to the apex location significantly decreased (p=0.019) the major thoracic Cobb angles.

Table 6.2 Statistical analyses of the major spinal curvatures of participants 1-6. The measurements through photogrammetry on the contralateral and ipsilateral sides relative to the apex under different loading weights were evaluated.

Asymmetric load (% BW)	**Loaded on the ipsilateral side of apex location**	**No-load condition**	**Loaded on the contralateral side of apex location**
	Pooled mean Cobb angle (degree) of the major curve		
0		17.4	
2.5	18.8		14.7
5	17.9		14.3
7.5	16.9		14.3
10	18.5		14.0
12.5	20.0		13.8
Repeated measure ANOVA	**P=0.773**		**p=0.019** **power=0.822**
Post hoc (LSD) pairwise comparison			**2.5% BW < 0% BW** **5% BW < 0% BW** **7.5% BW < 0% BW** **10% BW < 0% BW** **12.5% BW < 0% BW**

The mean major Cobb angle of the no-load condition (0% BW) was 17.4°. When the single-strap cross-chest bag was loaded at 2.5%, 5%, 7.5%, 10%, and 12.5% BW and positioned on the side of hip contralateral to the apex location, the mean major Cobb angles were 14.7°, 14.3°, 14.3°, 14.0°, and 13.8°, respectively. The mean major Cobb angle was significantly reduced under all loaded conditions comparing with no-load condition.

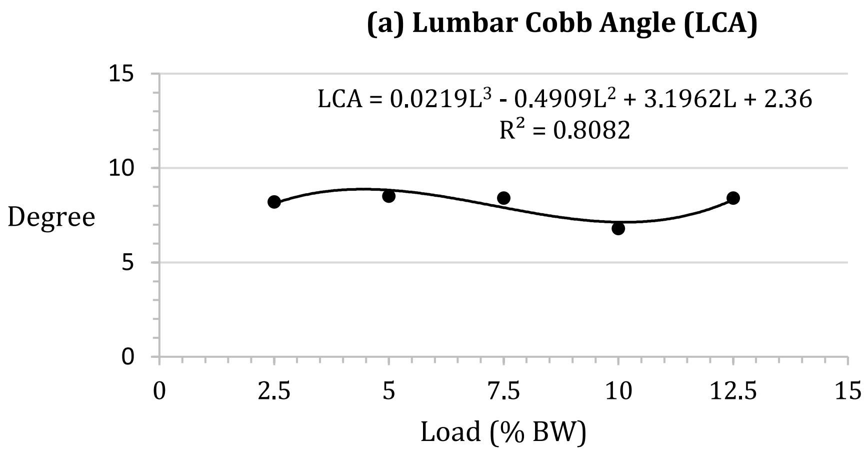
6.3.2 Loading on the effective side of apex location

The spinal curvatures of participant 7 were evaluated using the one-way ANOVA (Table 6.3). Under all loaded conditions on the effective side, the major thoracic Cobb angles were significantly reduced ($p<0.001$) but the minor lumbar Cobb angle did not change significantly ($p=0.413$). When the single-strap cross-chest bag was loaded at 2.5%, 5%, 7.5%, 10%, and 12.5% BW on the contralateral side, the mean major Cobb angles were 10.4°, 8.1°, 6.6°, 8.5°, and 8.1°, respectively. The mean major Cobb angle was significantly reduced under all loaded conditions comparing with no-load condition. The S-type scoliotic curve with major thoracic Cobb angle was corrected by placing the bag on the contralateral side of the location of the apex.

Table 6.3 Mean spinal curvatures (measured by photogrammetry) under no-load and loaded conditions on the contralateral side relative to the apex for participant 7.

Asymmetric load (% BW)	Mean Cobb Angle (degree)			
	Minor Lumbar Curve		Major Thoracic Curve	
	Mean	SD	Mean	SD
0	7.6	1.3	17.4	0.8
2.5	8.2	0.8	10.4	2.0
5	8.5	1.0	8.1	1.1
7.5	8.4	1.1	6.6	1.0
10	6.8	0.6	8.5	0.9
12.5	8.3	1.6	8.1	1.7
One-way ANOVA test	**p=0.431**		**p < 0.001 (power=1.000)**	
Post hoc (LSD) comparisons	**2.5%, 5%, 7.5%, 10%, 12.5% BW < 0% BW** **7.5% BW < 2.5% BW**			

A third order polynomial model was adopted to analyze the trend lines of scatter plots of the minor lumbar Cobb angle and major thoracic Cobb angle changes for participant 7 and evaluate the association between the spinal curvatures (minor lumbar Cobb angle and major thoracic Cobb angle) under various loaded conditions (Figure 6.2).



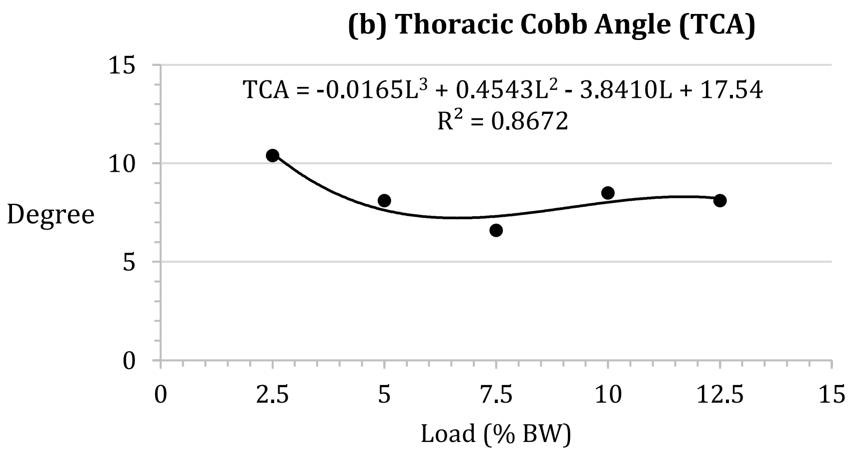


Figure 6.2 The third order polynomial regressions of the mean angles for participant 7 under each bag load. (a) Lumbar Cobb angle (LCA) and (b) Thoracic Cobb angle (TCA).

The LGP model for participant 7 was developed in accordance with the respective cost function (minimisation of the undesired deviational variables), constraint (formulations of the target levels), boundary (range of the decision variables), and non-negativity condition (Table 6.4). The optimal load was determined to be 4.1% of BW.

Table 6.4 LGP model formulated in accordance with the cost function, constraint, boundary, and nonnegativity conditions.

Lexicographic Goal Programming (LGP)	
Prioritized Goals: [1] Thoracic Cobb Angle (TCA), [2] Lumbar Cobb Angle (LCA)	
1st Goal Programming (GP)	
Cost Function	D^{+}_{TCA}
Constraint	$-0.0165L^3 + 0.4543L^2 - 3.8410L + D^{-}_{TCA} - D^{+}_{TCA} = -10.3050$ $0.0219L^3 - 0.4909L^2 + 3.1962L + D^{-}_{LCA} - D^{+}_{LCA} = 4.7680$
Boundary	$2.5 \le L \le 12.5$
Non-negativity	$D^{-}_{TCA}, D^{+}_{TCA}, D^{-}_{LCA}, D^{+}_{LCA} \ge 0$
2nd Goal Programming (GP)	
Cost Function	D^{+}_{LCA}
Constraint	$-0.0165L^3 + 0.4543L^2 - 3.8410L + D^{-}_{TCA} - D^{+}_{TCA} = -10.3050$ $0.0219L^3 - 0.4909L^2 + 3.1962L + D^{-}_{LCA} - D^{+}_{LCA} = 4.7680$ $D^{+}_{TCA}=1.0503$ [$D^{+}_{TCA}=1.0503$ was the solution of 1st GP]
Boundary	$2.5 \le L \le 12.5$
Nonnegativity	$D^{-}_{TCA}, D^{+}_{TCA}, D^{-}_{LCA}, D^{+}_{LCA} \ge 0$
Solution	**L = 4.1% BW**

6.4 Discussion

6.4.1 Spinal curvature measurements

The major thoracic Cobb angles of participants 1-6 were determined from a recent anterior-posterior X-ray film or clinical records issued by their doctors. The mean deviation of the spinal curvatures of participants 1-6, physically measured through photogrammetry of the X-ray films, was 1.6° (SD=0.7°). In a previous study applying similar photogrammetry approach, the mean deviation of the thoracic Cobb angle was 2.9° (Aroeira et al., 2011). The photogrammetry method used in this part of the study measured the thoracic Cobb with comparable accuracy.

6.4.2 Side effect of loading conditions

The mean spinal curvatures of all three participants with a right thoracic curve decreased when the load was on the left side of the hip (contralateral to the apex location), whereas those of the four with a left thoracic curve decreased when the load was on the right side of the hip (ipsilateral side of the apex location). Moreover, the spinal curvatures of participants 6 and 3 decreased (on average) when the load was on either side of the hip. However, the decreases in the spinal curvatures were smaller when the load was on the ipsilateral side relative to the apex location. These deviations might be due to the biomechanical effects including non-biomechanical impacts such as the fear of interference between the asymmetric load and the body individual variations in muscle strength activation in response to the application of external load (Zhang et al., 2010). When considering all the six participants, it was beneficial to place the shoulder bag contralateral to the apex location of the scoliotic curvature.

A previous study reported that the elevation of a single-loaded shoulder under asymmetric carriage generated lateral deviation of the trunk shift (Negrini and Negrini, 2007). This finding illustrated that the postural change under an asymmetric load was quite similar to side-shift exercises. However, because no muscle activation was evaluated in this study, further study is needed to illustrate the effects of asymmetric loading on muscle activity.

6.4.3 Optimal loading configuration

Although maximal decrease in the affected scoliotic region might improve coronal balance, it might also result in the compensation of sagittal alignment (Imrie et al., 2011). The positive decrease in the affected region might have negative impact on the unaffected regions of the spine. The R^2, proportion of the total variability of the experiment as explained by each regression model of the current data set, of the minor lumbar Cobb angle and major thoracic Cobb angle were 0.808 and 0.867, respectively. The two regression lines indicated moderately strong relationships between loading condition and spinal curvature. As the load increased, the minor lumbar Cobb angle increased from the baseline condition until it attained the first critical load at 4.8% of BW and then reduced until it reached the second critical load at 10.2% of BW (Figure 3.2a). By contrast, the major thoracic Cobb angle reduced from the baseline condition until it attained the first critical load at 6.6% of BW and then escalated until it reached the second critical load at 11.8% of BW (Figure 3.2b). Since the scoliotic spinal profile is subject dependent, the current regression models may exhibit continuous curvature changes under appropriate controlled asymmetric load carriage between 2.5% and 12.5% BW.

Imbalance in trunk muscle activation during asymmetric load carriage may escalate the shearing forces on the lumbar intervertebral discs and hence increase the risk of injury to these structures. Thereby, appropriate loading should be applied carefully to generate sufficient muscle activity for long-term spine correction without injuring the spine. The 4.1% of BW optimal load predicted in this study appears reasonable with decrease in the major thoracic curve by 9.1° and increase in the minor lumbar curve by only 1.1°.

6.4.4 Limitations

This part of the study has several limitations. First, only the short-term effects of asymmetric load carriage were investigated. Therefore, long-term effects should be evaluated in future studies. Second, young adults were the used in the preliminary study. Further study should be performed in children with scoliosis. Third, all participants possessed a major thoracic curve; hence, the effects on major scoliotic curves in other spinal regions may differ from the findings in this part of the study. Fourth, the effect of loads on the contralateral side of the apex, to reduce the major thoracic curve, might due to the Glenohumeral-shoulder rhythm motion.

6.5 Summary

A single-strap cross-chest bag provided short-term reduction of lateral curvature in young adults with scoliosis. A subject-specific optimal loading configuration could be determined using a multi-objective goal programming approach. Applying a controlled asymmetric load contralateral to the apex location of scoliosis successfully provided short-term postural correction and was a possible pragmatic method for correcting scoliotic spinal curvature. Further study on the long-term effects of subject-specific optimal asymmetric load carriage on spinal curvature in patients with scoliosis was warranted.

CHAPTER 7

Conclusion and future research

The study of this research project considered backpack and single-strap cross-chest bag carriage as an external perturbation on the spine and investigated the effects of symmetric backpack carriage for healthy individuals and asymmetric single-strap cross-strap bag carriage for patients with scoliosis in static and dynamic situations on the simultaneous changes in regional spinal curvature, trunk muscle activation, and lumbar spine loading as well as developed a refined biomechanical spine model for assessing lumbar spine loading using a multi-objective analysis approach.

7.1 Key findings and significant contributions

<u>Symmetric load carriage</u>

By assessing the simultaneous changes in regional spinal curvatures along the whole spine in the upright stance with backpack carriage up to a maximum of 20% of BW in young male adults, the most critical backpack load was found to be at 13% of BW. The most critical backpack load was evaluated with respect to the global critical load of the whole spine, instead of the local critical load of a specific spinal region or threshold level. A more comprehensive and reasonable standard considering both the postural changes and spinal stability as well as their associations with the increase in backpack load carriage could be determined.

By assessing the simultaneous changes in both trunk muscle activities and lumbar joint forces during walking with backpack carriage up to a maximum of 20% of BW in young male adults, the most critical backpack load was found to be at 10% of BW. The most critical backpack load was evaluated with respect to the critical load of the activities of six pairs of global trunk muscles and load-bearing strategies of three directional joint components at the lumbosacral level, instead of the local critical load of a specific trunk muscle or joint component force. Moreover, by assessing the lumbosacral joint force, lightweight backpack carriage at approximately 3% of BW might reduce the cyclic peak compression force by 3% of BW when compared with the no-load condition, and thus might help release the cyclic stress in the lumbar joint for patients with low back pain during walking. Moreover, the most critical changes in trunk muscle activation may be a good indicator for evaluating the load-bearing strategy in the lumbar spine during backpack carriage due to the lack of cocontraction strategy between the abdominal and back muscles.

By assessing the continuous changes in backpack load up to a maximum of 20% of BW and under a range of boundary condition of parametric muscle gain between 0.5 and 1.5 in the optimization process for predicting the lumbosacral joint compression force, a REMGAO approach was developed and the optimal parametric muscle gain was found to be converged at 0.98. The convergence of the parametric muscle gain close to 1 revealed that to maintain the stability of the trunk during walking, trunk muscle cocontraction should be sustained at a certain level, which is equal to the product of the normalized muscle EMG, physiological muscle cross-sectional area, and maximum muscle intensity.

<u>Asymmetric load carriage</u>

By assessing the simultaneous changes in both major scoliotic and minor scoliotic curves in the upright stance with asymmetric load carriage up to a maximum of 12.5% of BW in young adults with scoliosis, a lightweight (approximately 4% of BW in the preliminary study) single-strap cross-chest bag positioned at the hip level could provide short-term reduction of lateral curvature. Applying a controlled asymmetric load contralateral to the apex location of the major scoliotic curve successfully provided short-term postural correction and was a possible pragmatic method for correcting scoliotic spinal curvature.

<u>Multi-objective analysis approach</u>

By assessing the simultaneous changes in regional spinal curvature, trunk muscle activation, and lumbar spine loading during human activities, a protocol in multi-objective analysis approach was developed for determining the critical changes in physiological, kinematics, and kinetics responses to both symmetric and asymmetric load carriage in healthy individuals and patients in scoliosis, respectively.

7.2 Limitations and future research

Symmetric backpack load carriage studies

First, all participants were young male adults and the findings might not be generalizable to other gender or age groups. Second, all evaluations were short-term effect of backpack load carriage and the findings might not be applicable to long-term carriage. Third, the backpack load was set at a maximum of 20% of BW, which might not be applicable to load carriage beyond 20% of BW by hikers, firefighters, soldiers, or mountain porters.

Future studies involving female or in other age groups, in long-term carriage duration, and in load magnitude beyond 20% of BW in recreational hikers, or involving occupational or military activities are recommended to investigate the effects of backpack carriage covering the full spectrum of age groups and loading condition.

Asymmetric single-strap cross-chest bag load carriage study

First, young adult patients in scoliosis were recruited in the preliminary study. Second, only the short-term effects of asymmetric load carriage were investigated.

Third, all participants possessed a major thoracic curve; therefore, the effects on major scoliotic curves in other spinal regions might differ from the findings in the preliminary study. Fourth, the bag was located at the hip level.

Future studies in other age groups such as children, long-term carriage duration, and patients with different scoliotic profiles as well as location of the bag at other levels between the hip and trunk are recommended to further investigate the postural corrections of scoliotic curves in the coronal plane to generalize the results to other age groups and optimize the asymmetric load carriage effect on the correction of various scoliotic profile.

Multi-objective analysis spine model

The predictive optimal parametric gain of 0.98 of the REMGAO approach was based on the refinement of an EMGAO approach and a computation algorithm specifically for walking under various load conditions between no-load and a loaded backpack up to a maximum of 20% of BW. In addition, the comparability of the REMGAO approach was justified by the prediction of compression force in the lumbosacral joint. The REMGAO approach is simple in nature, efficient in computation, and unbiased in predicting the lumbosacral compression force of walking with a loaded backpack between 0% and 20% BW. The generalization of the REMGAO approach could be used in biomechanical modeling to estimate the mechanical loads of various human activities and joints during dynamic motions.

Appendix 1
Ethical approval, consent form, and information sheet for the study of backpack carriage

The Hong Kong Institute of Education
香港教育學院

Powering Education

8 May 2015

Professor CHOW Hung Kay Daniel
Head, Chair Professor
Department of Health and Physical Education

Dear Professor Chow,

Application for Ethical Review <Ref. no. 2014-2015-0316>

I am pleased to inform you that approval has been given by the Human Research Ethics Committee (HREC) for your research project:

Project title: Development of a Refined Biomechanical Model for Determining Lumbar Spine Loading

Ethical approval is granted for the project period from 8 May 2015 to 31 October 2016. If a project extension is applied for lasting more than 3 months, HREC should be contacted with information regarding the nature of and the reason for the extension. If any substantial changes have been made to the project, a new HREC application will be required.

Please note that you are responsible for informing the HREC in advance of any proposed substantive changes to the research proposal or procedures which may affect the validity of this ethical approval. You will receive separate notification should a fresh approval be required.

Thank you for your kind attention and we wish you well with your research.

Yours sincerely,

Connie Fung (Ms)
Secretary
Human Research Ethics Committee

c.c. Dr PARK Jae Hyung, Acting Chairperson, Human Research Ethics Committee

10 Lo Ping Road, Tai Po, New Territories, Hong Kong 香港新界大埔露屏路十號
T +852 2948 8888 F +852 2948 6000 www.ied.edu.hk

Appendix 1 (cont'd)
Ethical approval, consent form, and information sheet for the study of backpack carriage

<u>Consent Form and Information Sheet for PARTICIPANTS</u>

THE HONG KONG INSTITUTE OF EDUCATION

Department of Health & Physical Education

CONSENT TO PARTICIPATE IN RESEARCH

Development of a refined biomechanical model for determining lumbar spine loading

I ____________________ hereby consent to participate in the captioned research supervised by Prof. Daniel Chow and conducted by Simon Li.

I understand that information obtained from this research may be used in future research and may be published. However, my right to privacy will be retained, i.e., my personal details will not be revealed.

The procedure as set out in the **<u>attached</u>** information sheet has been fully explained. I understand the benefits and risks involved. My participation in the project is voluntary.

I acknowledge that I have the right to question any part of the procedure and can withdraw at any time without negative consequences.

Name of participant ______________________________

Signature of participant ______________________________

Name of parent/guidance ______________________________

Signature of parent/guidance ______________________________

Date ______________________________

Appendix 1 (cont'd)
Ethical approval, consent form, and information sheet for the study of backpack carriage

INFORMATION SHEET

Development of a refined biomechanical model for determining lumbar spine loading

You are invited to participate in a project supervised by Prof. Daniel Chow and conducted by Simon Li. They are staff of the Department of Health & Physical Education in The Hong Kong Institute of Education.

Low back disorders (LBDs) and low back pain (LBP) have been the vital health problem all over the world. They are either acute injuries, chronic illnesses or perceived pain that are generated during improper workplace material handling tasks, physically demanding jobs or load carriage activities. Two researches done in Hong Kong and United States validated that occupational LBDs and LBP have had significant long term economic impact on all sectors of the society. They lengthen sick leave period, reduce productivity, increase cost, and consequently consume resources. The preventive measures of LBDs and LBP are critical especially for the most injury part of the spine at the lumbar joints. While the direct invasive assessment of the realistic loading on the lumbar spine is inappropriate, the indirect non-invasive approach of modeling is commonly used to approximate the muscle forces and moments acting on the lumbar spine during human dynamic motions.

Previous studies of biomechanical models in predicting the loadings on the lumbar spine had made necessary assumptions regarding certain conditions under consideration as well as the limitations of work place analysis. These assumptions had limited their accuracy in approximating the realistic free dynamic loading situations. The aims of this study are to construct a refined electromyography assisted optimization (REMGAO) model to approximate, with comparable accuracy, lumbar spine loading in terms of compression, anterior-posterior and lateral shear forces and validate the possibility of generalizing the REMGAO model to predict the loading of individual joint in accordance with respective group of muscles.

A REMGAO model will be developed and validated by a serious of static, dynamic or free dynamic activities including posture control, material lifting and level walking. You will be instructed to perform four tasks: (1) erect stance without or with 4 critical backpack loads, (2) lifting a load of 116N in front at knee level, (3) level walking without and with 4 critical backpack loads and (4) maximum voluntary contraction activities in 4 postures (back lift pull, shoulder extension, trunk extension and bent sit up). You will perform the tasks on or walk along a pre-set walkway across a force platform and three trials for each activity. You will be

Appendix 1 (cont'd)
Ethical approval, consent form, and information sheet for the study of backpack carriage

allowed to rest for 1 minute to 2 minutes between each trial and 5 minutes after completing a set of task.

The complete experiment will last for about two hours. Transportation allowance (HKD 200) will be provided for each visit.

Anthropometry data of your lower body and limbs will be measured. Then, you will be trained to get familiar with all tasks prior to the experiment. Twelve pairs of electrodes and five inertia measuring units will be attached at your upper body to capture the electromyography activities of various muscles and spinal curvatures. Twenty-five reflective markers will be attached at your lower limbs and upper body to capture the motion of the body segments.

You may experience minor discomfort during material lifting and backpack load carriage and your spinal deformity may transiently be increased or decreased. You will not benefit from the experiment and the results of the study will be used for developing future treatments for subjects with scoliosis. You have every right to withdraw from the study at any time without negative consequences. All information related to you will remain confidential, and will be identifiable by codes known only to the researcher.

If you have any concerns about the conduct of this research study, please do not hesitate to contact the Human Research Ethics Committee by email at hrec@ied.edu.hk or by mail to Research and Development Office, The Hong Kong Institute of Education (Tel: 2948-6318).

If you would like to obtain more information about this study, please contact Prof. Daniel Chow at telephone number 2948- .

Thank you for your interest in participating in this study.

Prof. Daniel Chow
Principal Investigator

Appendix 2
Ethical approval, consent form, and information sheet for the study of single-strap cross-chest bag carriage

The Hong Kong Institute of Education
香港教育學院

Powering Education

27 March 2015

Professor CHOW Hung Kay Daniel
Chair Professor
Department of Health and Physical Education

Dear Professor Chow,

Application for Ethical Review <Ref. no. 2014-2015-0130>

I am pleased to inform you that approval has been given by the Human Research Ethics Committee (HREC) for your research project:

Project title: Long-term Effect of Asymmetric Loading on Spinal Deformity in Adolescent Idiopathic Scoliosis - A Feasible Study

Ethical approval is granted for the project period from 1 July 2015 to 30 June 2018. If a project extension is applied for lasting more than 3 months, HREC should be contacted with information regarding the nature of and the reason for the extension. If any substantial changes have been made to the project, a new HREC application will be required.

Please note that you are responsible for informing the HREC in advance of any proposed substantive changes to the research proposal or procedures which may affect the validity of this ethical approval. You will receive separate notification should a fresh approval be required.

Thank you for your kind attention and we wish you well with your research.

Yours sincerely,

Connie Fung (Ms)
Secretary
Human Research Ethics Committee

c.c. Dr PARK Jae Hyung, Acting Chairperson, Human Research Ethics Committee

10 Lo Ping Road, Tai Po, New Territories, Hong Kong 香港新界大埔露屏路十號
T +852 2948 8888 F +852 2948 6000 www.ied.edu.hk

Appendix 2 (cont'd)
Ethical approval, consent form, and information sheet for the study of single-strap cross-chest bag carriage

Consent Form and Information Sheet for PARTICIPANTS

THE HONG KONG INSTITUTE OF EDUCATION

Department of Health & Physical Education

CONSENT TO PARTICIPATE IN RESEARCH

Long-term Effect of Asymmetric Loading on Spinal Deformity in Adolescent Idiopathis Scoliosis – A Feasible Study

I _______________ hereby consent to participate in the captioned research supervised by Prof. Daniel Chow and conducted by Simon Li.

I understand that information obtained from this research may be used in future research and may be published. However, my right to privacy will be retained, i.e., my personal details will not be revealed.

The procedure as set out in the **attached** information sheet has been fully explained. I understand the benefits and risks involved. My participation in the project is voluntary.

I acknowledge that I have the right to question any part of the procedure and can withdraw at any time without negative consequences.

Name of participant ______________________________

Signature of participant ______________________________

Name of parent/guidance ______________________________

Signature of parent/guidance ______________________________

Date ______________________________

Appendix 2 (cont'd)
Ethical approval, consent form, and information sheet for the study of single-strap cross-chest bag carriage

INFORMATION SHEET

Long-term Effect of Asymmetric Loading on Spinal Deformity in Adolescent Idiopathis Scoliosis – A Feasible Study

You are invited to participate in a project supervised by Prof. Daniel Chow and conducted by Simon Li. They are staff of the Department of Health & Physical Education in The Hong Kong Institute of Education.

Scoliosis is a three-dimensional deformity of the spine which typically affects 4-6% of growing adolescents. As this spinal deformity was demonstrated to be associated with poorer pulmonary function and higher prevalence of back pain, as well as motor control impairment and psychosocial problem, therapeutic management should be prescribed once the deformity is identified. The major goal of the management is to prevent the curvature from progression. Physiotherapy exercise is usually prescribed as the preliminary treatment for individuals with adolescent idiopathic scoliosis. It was demonstrated to be clinically effective in reducing the curve progression rate and bracing prescription with improved mobility and balance. The effects were found to be particularly positive for side-shift and asymmetric exercises. However, similar to the application of bracing treatment, compliance with the prescribed exercise protocol is difficult as it is practically not feasible to request children to attend twice or thrice training sessions per week or to perform self-administered regular exercises by themselves at home. It would be a pragmatic approach if these regular exercises could be integrated with children's daily activities. Load carriage is common in school-age children for transferring books and personal belongings to and from schools. It is conventionally believed that abnormal external loading is one of the possible factors that may exacerbate spinal deformity. Thus, children are usually recommended to carry the load symmetrically over the shoulders. However, as asymmetric and side-shift exercises have been demonstrated to be effective therapeutic exercises for scoliosis management, we propose the use of properly controlled asymmetric load for postural correction and muscle conditioning in adolescents with idiopathic scoliosis. In this study, the biomechanical effects of asymmetric load carriage of different weights on spinal curvature, body posture and muscle activation patterns will be investigated in adolescents with idiopathic scoliosis. The findings of the study will contribute knowledge to the establishment of guidelines for incorporating asymmetric load carriage into the daily routine of adolescents with idiopathic scoliosis for postural correction and muscle conditioning.

Appendix 2 (cont'd)
Ethical approval, consent form, and information sheet for the study of single-strap cross-chest bag carriage

You will need to participate in the study twice. Each visit will last for about 2 hours. Transportation allowance (HKD 200) will be provided for each visit. The objective of the first visit is to determine the optimum position and weight of the asymmetric load that can achieve the maximum correction of spinal curvature. You will be instructed to carry a shoulder bag with the weight increasing incrementally at 5N (i.e. 0.5kg) up to the maximum of 12.5% of your body weight. The spinal curvature and standing posture will be measured non-invasively using an ultrasound system and laser plumb lines, respectively. In the second visit, the optimal asymmetric loading configuration identified in the first visit will be employed for evaluating the muscle training effects. The participants will be invited to come to the laboratory in about a week time. The activity of the participants' trunk muscles will be collected using a non-invasive electromyography (EMG) system during (1) relaxed upright standing, (2) side-shift exercise, (3) asymmetric exercise, and (4) optimal asymmetric load carriage.

Participants will be trained by an experienced physiotherapist to get familiar with the asymmetric and side-shift exercises prior to the experiment. EMG activities of various muscles will be captured during relaxed upright standing as the baseline. The order of EMG analysis for participants under conventional therapeutic exercises and optimal asymmetric load carriage will be randomized. Three measurement trials will be performed and the results will be averaged for data analysis. The subjects will be allowed to rest for at least 5 minutes prior to each testing condition.

You may experience minor discomfort during short-term asymmetric load carriage and your spinal deformity may transiently be increased or decreased. You will not benefit from the experiment and the results of the study will be used for developing future treatments for subjects with scoliosis. You have every right to withdraw from the study at any time without negative consequences. All information related to you will remain confidential, and will be identifiable by codes known only to the researcher.

If you have any concerns about the conduct of this research study, please do not hesitate to contact the Human Research Ethics Committee by email at hrec@ied.edu.hk or by mail to Research and Development Office, The Hong Kong Institute of Education (Tel: 2948-6318).

Appendix 2 (cont'd)
Ethical approval, consent form, and information sheet for the study of single-strap cross-chest bag carriage

If you would like to obtain more information about this study, please contact Prof. Daniel Chow at telephone number 2948 .

Thank you for your interest in participating in this study.

Prof. Daniel Chow
Principal Investigator

Appendix 3.1
Experimental data of sagittal spinal curvatures under backpack carriage in upright stance

Appendix 3.1.1 Head on Neck Lordosis

Load (% BW)	0			5			10			15			20		
Trial	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**
Participant 1	44.4	44.1	45.0	47.2	43.3	46.9	46.6	46.1	46.9	46.8	45.8	45.8	43.0	43.6	43.9
Participant 2	21.6	22.0	22.2	24.1	21.0	23.2	22.8	21.5	24.9	26.8	24.3	22.6	21.3	17.7	15.9
Participant 3	36.5	37.1	37.8	41.3	39.9	41.0	38.3	39.9	39.1	35.8	36.6	39.7	42.6	38.2	41.0
Participant 4	38.3	37.2	38.9	42.6	45.7	43.7	43.4	44.6	45.3	41.8	40.5	42.9	42.4	43.4	42.0
Participant 5	37.2	39.0	38.1	40.3	46.1	40.3	43.2	42.9	39.9	45.5	43.0	43.2	45.9	46.3	42.6
Participant 6	32.7	29.8	31.2	26.5	27.6	28.8	26.0	25.3	27.4	26.3	27.4	25.8	25.7	23.8	26.7
Participant 7	25.1	25.8	23.8	28.3	28.2	28.5	31.8	29.1	25.4	28.8	27.8	26.3	30.2	31.4	29.0
Participant 8	21.3	21.4	20.7	20.1	19.2	17.9	19.8	22.2	19.7	22.5	23.1	22.6	14.9	26.0	18.2
Participant 9	17.7	18.1	18.2	22.2	24.1	21.4	24.0	22.3	23.4	22.3	21.8	22.8	18.9	19.8	17.9
Participant 10	46.2	45.1	47.1	43.5	42.5	43.0	43.6	44.5	42.7	45.2	43.9	42.7	42.6	45.7	41.2
Mean	32.1	32.0	32.3	33.6	33.7	33.5	34.0	33.8	33.5	34.2	33.4	33.5	32.8	33.6	31.8
SD	10.1	9.8	10.5	10.3	10.7	10.7	10.2	10.6	10.3	10.0	9.5	10.1	11.8	11.2	11.6
Overall mean	**32.1**			**33.6**			**33.8**			**33.7**			**32.7**		
Overall SD	**9.8**			**10.2**			**10.0**			**9.5**			**11.2**		

Appendix 3.1.2 Upper Thoracic Kyphosis

Load (% BW)	0			5			10			15			20		
Trial	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**
Participant 1	41.1	38.9	40.9	34.0	35.3	34.5	26.5	27.0	29.0	29.1	30.1	28.4	29.6	29.3	30.7
Participant 2	19.4	19.7	20.5	20.5	19.0	21.7	19.3	18.7	20.7	16.3	18.2	18.9	19.1	22.1	23.1
Participant 3	46.4	44.3	45.2	37.6	35.4	39.4	37.5	39.0	38.2	37.6	35.5	40.1	43.4	38.5	39.3
Participant 4	40.4	39.3	38.4	36.5	37.4	33.7	38.0	35.3	34.0	37.9	36.6	35.0	36.9	36.4	34.5
Participant 5	38.4	38.1	38.5	31.5	30.6	31.5	28.8	27.3	28.2	28.1	27.2	26.6	26.2	27.9	26.7
Participant 6	32.1	33.9	33.1	32.6	29.0	28.6	29.1	31.3	31.8	30.9	29.9	26.1	28.7	26.9	29.3
Participant 7	20.7	26.3	24.1	20.8	20.4	21.1	16.3	15.4	18.5	17.1	19.2	19.0	15.8	12.4	15.5
Participant 8	29.5	29.9	29.8	26.0	34.9	33.9	25.8	31.0	29.8	26.5	28.6	32.2	29.0	26.8	25.7
Participant 9	16.8	17.6	17.2	9.8	8.3	9.4	8.8	11.2	8.5	8.8	8.4	6.4	14.9	15.7	14.1
Participant 10	42.4	41.7	42.2	40.9	40.0	40.1	41.0	41.1	39.9	39.6	40.7	40.5	41.1	37.8	40.2
Mean	32.7	33.0	33.0	29.0	29.0	29.4	27.1	27.7	27.9	27.2	27.4	27.3	28.5	27.4	27.9
SD	10.7	9.3	9.7	9.6	10.1	9.5	10.2	10.0	9.6	10.3	9.8	10.5	10.0	8.9	8.9
Overall mean	**32.9**			**29.1**			**27.6**			**27.3**			**27.9**		
Overall SD	**9.6**			**9.4**			**9.6**			**9.9**			**8.9**		

Appendix 3.1.3 Lower Thoracic Kyphosis

Load (% BW)	0			5			10			15			20		
Trial	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**
Participant 1	5.5	5.6	5.6	7.0	6.0	6.2	9.0	10.3	8.6	8.1	8.2	8.7	8.0	9.3	7.6
Participant 2	1.6	1.3	1.5	4.1	4.1	4.0	4.1	4.3	4.4	3.1	3.2	3.5	3.0	3.4	2.7
Participant 3	8.5	9.9	10.2	15.1	15.7	16.3	10.3	13.3	11.4	12.2	13.3	12.5	12.4	15.1	14.3
Participant 4	13.1	11.3	12.2	16.2	20.3	19.7	17.5	20.0	20.6	18.5	18.0	18.7	17.1	16.8	18.1
Participant 5	6.8	7.4	6.1	6.2	7.5	6.2	3.6	5.3	1.8	7.7	6.2	7.1	5.0	5.9	4.5
Participant 6	25.7	26.6	24.5	25.1	27.5	27.3	29.7	28.4	26.6	29.3	27.3	26.6	29.5	30.4	29.3
Participant 7	11.8	10.7	9.6	11.9	12.1	11.7	12.3	13.5	15.5	11.9	10.2	14.1	8.1	13.0	11.1
Participant 8	12.4	11.6	11.7	10.6	9.4	8.5	10.1	10.2	12.0	11.8	12.1	12.0	10.0	12.1	11.5
Participant 9	15.1	13.7	14.7	18.0	17.7	18.5	18.5	21.2	17.6	17.6	16.3	16.8	14.3	15.6	15.0
Participant 10	3.8	5.2	4.7	5.1	5.9	4.0	3.7	4.2	3.3	6.4	4.4	4.5	5.8	4.0	4.2
Mean	10.4	10.3	10.1	11.9	12.6	12.2	11.9	13.1	12.2	12.7	11.9	12.5	11.3	12.6	11.8
SD	6.9	6.8	6.5	6.7	7.5	7.9	8.2	8.0	8.0	7.5	7.3	7.1	7.7	7.9	8.0
Overall mean	**10.3**			**12.3**			**12.4**			**12.3**			**11.9**		
Overall SD	**6.5**			**7.1**			**7.8**			**7.0**			**7.6**		

Appendix 3.1.4 Lumbar Lordosis

Load (% BW)	0			5			10			15			20		
Trial	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**	**1**	**2**	**3**
Participant 1	20.3	19.9	20.2	18.7	19.7	19.6	16.7	15.4	17.1	18.4	18.0	18.6	17.7	16.4	18.1
Participant 2	18.1	17.2	17.6	13.6	16.8	14.3	17.7	16.6	18.7	15.4	15.3	16.6	16.3	16.6	16.4
Participant 3	17.8	19.1	18.2	13.0	14.6	17.0	19.2	15.0	17.1	19.0	18.2	17.2	22.5	19.1	15.6
Participant 4	15.4	16.6	14.8	10.8	13.4	14.0	10.6	12.3	11.9	10.4	9.5	11.2	12.8	10.4	10.3
Participant 5	9.3	9.1	9.2	6.6	7.7	6.6	4.4	10.5	9.9	11.0	10.4	5.0	6.0	6.5	7.2
Participant 6	22.0	23.1	22.4	19.9	18.3	18.1	16.1	15.8	17.5	18.6	18.8	18.2	16.0	14.6	18.2
Participant 7	14.9	14.6	14.7	11.6	11.6	11.7	12.1	16.0	14.5	12.6	12.2	10.7	13.1	14.8	13.4
Participant 8	24.6	25.6	25.6	22.7	30.1	28.5	23.5	31.4	26.8	22.3	30.4	26.8	18.9	26.0	21.9
Participant 9	16.6	17.6	17.1	16.4	16.7	16.9	15.7	15.9	11.8	14.2	13.5	13.4	15.4	16.6	14.1
Participant 10	11.1	11.9	13.0	11.8	10.9	9.7	9.3	7.7	9.9	8.8	7.3	10.4	13.5	14.6	12.2
Mean	17.0	17.5	17.3	14.5	16.0	15.6	14.5	15.7	15.5	15.1	15.4	14.8	15.2	15.6	14.7
SD	4.7	4.9	4.7	4.9	6.2	6.0	5.5	6.2	5.1	4.4	6.6	6.0	4.4	5.1	4.3
Overall mean	**17.3**			**15.4**			**15.2**			**15.1**			**15.2**		
Overall SD	**4.6**			**5.6**			**5.5**			**5.5**			**4.5**		

Appendix 4.1
Experimental data of spatial-temporal parameters under backpack carriage during walking

Appendix 4.1.1 Cadence (steps/min)

Participant	Load (% BW)				
	0	5	10	15	20
1	117.8	115.2	119.9	102.8	109.3
2	121.4	120.3	116.5	118.7	117.6
3	118.4	116.9	114.8	116.5	110.1
4	115.6	117.5	118.5	117.8	112.9
5	107.7	110.6	108.8	107.9	107.6
6	123.4	120.3	111.1	117.5	122.0
7	123.2	123.2	121.1	122.7	117.8
8	109.0	124.1	119.8	120.4	119.8
9	134.1	120.9	122.1	119.5	121.4
10	129.4	131.9	131.5	131.8	132.8
Mean	**120.0**	**120.1**	**118.4**	**117.6**	**117.1**
SD	**8.2**	**5.7**	**6.3**	**7.9**	**7.6**

Appendix 4.1.2 Walking speed (m/s)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	1.32	1.30	1.34	1.14	1.24
2	1.35	1.34	1.29	1.32	1.30
3	1.20	1.23	1.20	1.21	1.14
4	1.24	1.26	1.24	1.26	1.19
5	1.21	1.27	1.28	1.23	1.22
6	1.19	1.18	1.03	1.14	1.17
7	1.28	1.28	1.26	1.31	1.20
8	1.17	1.38	1.29	1.37	1.28
9	1.50	1.32	1.31	1.28	1.35
10	1.39	1.46	1.45	1.44	1.42
Mean	**1.29**	**1.30**	**1.27**	**1.27**	**1.25**
SD	**0.11**	**0.08**	**0.11**	**0.10**	**0.09**

Appendix 4.1.3 Stride length (m)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	1.34	1.35	1.34	1.33	1.36
2	1.33	1.34	1.33	1.33	1.33
3	1.21	1.26	1.26	1.25	1.25
4	1.29	1.29	1.26	1.28	1.26
5	1.35	1.38	1.41	1.37	1.36
6	1.16	1.18	1.12	1.17	1.15
7	1.24	1.25	1.25	1.28	1.23
8	1.28	1.34	1.30	1.36	1.28
9	1.34	1.31	1.28	1.29	1.33
10	1.29	1.32	1.32	1.31	1.28
Mean	**1.28**	**1.30**	**1.29**	**1.30**	**1.28**
SD	**0.06**	**0.06**	**0.08**	**0.06**	**0.07**

Appendix 4.1.4 Stride time (s)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	1.02	1.04	1.00	1.17	1.10
2	0.99	1.00	1.03	1.01	1.02
3	1.01	1.03	1.05	1.03	1.09
4	1.04	1.02	1.01	1.02	1.06
5	1.11	1.09	1.10	1.11	1.12
6	0.97	1.00	1.08	1.02	0.98
7	0.97	0.97	0.99	0.98	1.02
8	1.10	0.97	1.00	1.00	1.00
9	0.89	0.99	0.98	1.00	0.99
10	0.93	0.91	0.91	0.91	0.90
Mean	**1.00**	**1.00**	**1.02**	**1.03**	**1.03**
SD	**0.07**	**0.05**	**0.05**	**0.07**	**0.07**

Appendix 4.1.5 Step length (m)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	0.74	0.71	0.62	0.71	0.74
2	0.66	0.64	0.63	0.62	0.59
3	0.63	0.63	0.61	0.63	0.63
4	0.64	0.63	0.63	0.60	0.63
5	0.64	0.67	0.67	0.64	0.63
6	0.58	0.58	0.58	0.56	0.58
7	0.63	0.63	0.63	0.62	0.62
8	0.66	0.70	0.67	0.69	0.67
9	0.66	0.66	0.66	0.65	0.64
10	0.67	0.68	0.68	0.67	0.67
Mean	**0.65**	**0.65**	**0.64**	**0.64**	**0.64**
SD	**0.04**	**0.04**	**0.03**	**0.04**	**0.05**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	0.61	0.64	0.72	0.63	0.63
2	0.68	0.70	0.70	0.72	0.74
3	0.58	0.64	0.65	0.62	0.62
4	0.66	0.66	0.62	0.68	0.63
5	0.71	0.71	0.74	0.73	0.73
6	0.58	0.60	0.53	0.61	0.57
7	0.62	0.62	0.62	0.66	0.61
8	0.63	0.63	0.62	0.68	0.61
9	0.69	0.66	0.63	0.64	0.69
10	0.62	0.65	0.64	0.64	0.61
Mean	**0.64**	**0.65**	**0.65**	**0.66**	**0.64**
SD	**0.05**	**0.03**	**0.06**	**0.04**	**0.06**

Appendix 4.1.6 Step time (s)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	0.55	0.57	0.55	0.62	0.58
2	0.51	0.52	0.53	0.52	0.53
3	0.43	0.43	0.44	0.42	0.47
4	0.54	0.54	0.53	0.53	0.57
5	0.56	0.58	0.58	0.56	0.58
6	0.54	0.55	0.56	0.54	0.53
7	0.44	0.42	0.44	0.42	0.45
8	0.52	0.49	0.51	0.51	0.52
9	0.48	0.49	0.49	0.49	0.49
10	0.49	0.50	0.49	0.49	0.49
Mean	**0.51**	**0.51**	**0.51**	**0.51**	**0.52**
SD	**0.04**	**0.05**	**0.05**	**0.06**	**0.05**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	0.47	0.48	0.45	0.54	0.52
2	0.48	0.48	0.50	0.50	0.49
3	0.58	0.60	0.61	0.61	0.62
4	0.49	0.49	0.49	0.49	0.49
5	0.56	0.51	0.52	0.55	0.54
6	0.43	0.45	0.52	0.48	0.46
7	0.54	0.55	0.55	0.56	0.57
8	0.58	0.48	0.49	0.49	0.49
9	0.41	0.50	0.50	0.51	0.50
10	0.43	0.41	0.42	0.42	0.42
Mean	**0.50**	**0.50**	**0.51**	**0.52**	**0.51**
SD	**0.06**	**0.05**	**0.05**	**0.05**	**0.05**

Appendix 4.1.7 Single support duration (% gait cycle)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	41.6	36.3	41.7	40.5	41.3
2	39.0	38.5	40.1	39.3	40.2
3	39.6	36.8	38.2	37.5	39.4
4	43.9	37.1	42.4	41.5	50.0
5	38.5	35.8	40.4	38.4	38.2
6	40.2	39.9	37.2	38.2	38.8
7	39.0	39.4	37.0	38.5	36.4
8	35.8	40.3	36.4	36.4	36.3
9	38.1	39.6	38.6	38.2	37.0
10	41.1	41.5	41.0	40.3	39.6
Mean	**39.7**	**38.5**	**39.3**	**38.9**	**39.7**
SD	**2.2**	**1.9**	**2.1**	**1.5**	**4.0**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	37.7	42.7	32.3	36.4	37.2
2	39.2	39.9	38.3	37.9	37.6
3	37.7	40.6	36.6	36.8	35.5
4	37.9	43.3	38.1	36.8	36.5
5	38.7	41.7	35.8	36.2	35.0
6	40.9	40.7	38.0	38.4	37.9
7	40.6	39.4	39.1	38.6	38.1
8	38.0	35.3	39.5	39.7	39.2
9	40.0	38.8	40.2	37.8	38.1
10	40.3	40.5	41.4	40.8	41.1
Mean	**39.1**	**40.3**	**37.9**	**37.9**	**37.6**
SD	**1.3**	**2.2**	**2.6**	**1.5**	**1.8**

Appendix 4.1.8 Double support duration (% gait cycle)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	10.3	10.5	13.0	11.5	10.7
2	10.9	10.8	10.8	11.4	11.1
3	11.3	11.3	12.6	12.9	12.6
4	9.1	9.8	9.7	10.9	6.8
5	11.4	11.2	11.9	12.7	13.4
6	9.5	9.7	12.4	11.7	11.7
7	10.2	10.6	11.9	11.4	12.8
8	13.1	12.2	12.1	12.0	12.3
9	11.0	10.8	10.6	12.0	12.5
10	9.3	9.0	8.8	9.5	9.6
Mean	**10.6**	**10.6**	**11.4**	**11.6**	**11.4**
SD	**1.2**	**0.9**	**1.4**	**1.0**	**2.0**

Appendix 4.2
Experimental data of ground reaction forces under backpack carriage during walking

Appendix 4.2.1 First peak of vertical force (% BW)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	118.9	129.5	128.0	131.2	141.2
2	113.1	113.9	118.9	123.0	127.3
3	102.1	115.7	115.0	113.9	123.1
4	107.8	108.2	117.9	123.8	123.6
5	116.1	120.4	125.2	127.8	133.9
6	102.7	109.6	111.3	109.6	120.3
7	112.1	117.8	122.8	136.0	130.0
8	115.8	121.8	124.3	133.4	134.1
9	121.7	126.3	127.6	132.6	137.8
10	97.6	113.6	108.9	109.2	120.2
Mean	**110.8**	**117.7**	**120.0**	**124.1**	**129.2**
SD	**8.0**	**6.9**	**6.7**	**10.0**	**7.4**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	119.1	116.4	119.7	123.5	133.2
2	111.9	121.2	129.8	129.8	126.7
3	114.4	122.3	119.2	129.0	131.1
4	111.4	115.2	121.6	127.5	127.7
5	110.7	115.0	119.6	121.5	131.4
6	105.8	111.0	113.3	114.0	125.2
7	114.3	123.0	124.0	132.2	138.7
8	118.5	121.1	123.3	128.7	133.2
9	124.8	127.8	131.4	139.8	148.5
10	96.4	103.7	108.1	108.8	119.0
Mean	**112.7**	**117.7**	**121.0**	**125.5**	**131.5**
SD	**7.8**	**6.9**	**6.9**	**9.0**	**8.0**

Appendix 4.2.2 Second peak of vertical force (% BW)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	105.0	114.0	118.2	120.2	123.7
2	113.1	113.9	118.9	123.0	127.3
3	102.1	115.7	115.0	113.9	123.1
4	110.5	120.5	127.9	132.7	128.3
5	116.1	120.4	125.2	127.8	133.9
6	110.2	119.5	122.6	128.7	136.8
7	101.5	112.3	118.4	121.1	126.7
8	108.8	114.2	120.7	131.4	129.8
9	113.0	119.8	120.5	125.7	130.9
10	116.5	108.8	132.1	129.5	137.2
Mean	**109.7**	**115.9**	**122.0**	**125.4**	**129.8**
SD	**5.4**	**4.0**	**5.1**	**5.8**	**5.0**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	100.6	103.9	110.8	114.6	121.7
2	105.7	108.2	117.8	122.7	124.1
3	105.5	120.1	112.8	119.9	128.1
4	110.3	117.2	123.6	127.7	124.3
5	114.4	117.4	124.6	132.4	136.1
6	112.5	119.9	124.8	131.6	135.7
7	102.2	117.0	115.1	125.0	135.5
8	109.1	117.5	117.7	122.0	131.2
9	111.7	118.1	127.1	128.5	131.7
10	114.4	97.4	125.6	129.5	134.5
Mean	**108.6**	**113.7**	**120.0**	**125.4**	**130.3**
SD	4.9	7.8	5.9	5.6	5.4

Appendix 4.2.3 Trough of vertical force (% BW)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	69.7	70.0	75.6	83.5	85.8
2	65.8	71.0	71.9	80.1	83.9
3	72.8	75.3	78.8	83.3	90.1
4	73.7	80.2	76.8	88.4	88.7
5	73.0	79.9	85.2	88.0	91.5
6	82.2	83.4	89.6	92.7	91.8
7	74.8	74.6	79.0	79.7	91.6
8	78.1	78.9	85.5	84.6	93.3
9	65.7	68.4	79.3	81.1	75.5
10	76.3	53.6	87.0	83.4	91.9
Mean	**73.2**	**73.5**	**80.9**	**84.5**	**88.4**
SD	**5.2**	**8.6**	**5.7**	**4.1**	**5.4**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	71.4	79.6	80.6	89.1	86.2
2	63.9	68.4	78.4	84.4	78.4
3	71.5	67.7	79.4	80.3	85.5
4	73.0	75.9	77.1	88.4	92.7
5	73.7	77.6	82.4	86.0	88.0
6	79.9	85.6	58.7	91.4	93.0
7	71.9	67.3	80.4	82.2	86.5
8	71.7	72.3	79.2	75.2	84.7
9	63.9	70.0	77.7	80.2	79.3
10	71.8	51.8	81.5	80.5	89.5
Mean	**71.3**	**71.6**	**77.5**	**83.8**	**86.4**
SD	**4.6**	**9.2**	**6.8**	**5.0**	**4.9**

Appendix 4.2.4 Peak of anterior shear force (% BW)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	22.4	21.3	18.5	23.5	25.4
2	25.3	24.7	25.6	26.1	28.8
3	26.2	26.1	27.0	29.5	26.8
4	21.3	22.3	24.1	24.7	24.2
5	26.5	26.0	24.3	27.2	28.8
6	19.1	21.3	21.1	19.4	22.5
7	21.5	25.3	27.9	26.2	28.2
8	22.9	26.8	28.5	31.7	31.0
9	23.0	25.1	27.2	26.8	28.9
10	21.6	23.6	22.7	24.3	26.1
Mean	**23.0**	**24.3**	**24.7**	**25.9**	**27.1**
SD	**2.4**	**2.0**	**3.2**	**3.4**	**2.6**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	20.8	20.5	21.5	23.7	24.8
2	27.5	28.1	26.0	26.8	30.2
3	21.3	23.6	22.7	26.4	23.6
4	21.1	23.3	27.1	25.4	27.2
5	21.5	23.7	24.3	23.4	26.7
6	16.6	18.9	18.8	19.2	20.1
7	19.8	22.5	24.1	24.3	24.7
8	32.7	34.4	34.3	36.0	35.5
9	29.8	28.9	28.4	28.5	34.4
10	24.1	23.6	27.3	26.7	27.9
Mean	**23.5**	**24.8**	**25.5**	**26.0**	**27.5**
SD	**5.0**	**4.5**	**4.3**	**4.3**	**4.8**

Appendix 4.2.5 Peak of posterior shear force (% BW)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	24.0	25.0	23.3	27.0	28.7
2	27.3	28.8	25.8	26.2	28.7
3	19.8	23.2	19.5	22.1	22.3
4	20.2	19.8	23.3	23.6	22.1
5	16.1	19.5	22.1	19.9	23.6
6	14.3	16.5	15.4	16.5	20.3
7	17.7	22.1	20.6	21.6	23.2
8	16.6	21.5	21.7	21.5	23.8
9	19.0	19.4	24.1	25.2	24.3
10	24.7	22.3	23.6	27.3	24.6
Mean	**20.0**	**21.8**	**21.9**	**23.1**	**24.2**
SD	**4.2**	**3.4**	**2.9**	**3.4**	**2.7**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	27.6	27.8	23.2	28.2	31.5
2	25.7	23.8	26.9	24.6	29.8
3	18.4	23.3	21.0	23.1	21.3
4	20.6	18.1	21.4	21.5	22.3
5	17.6	18.6	19.6	19.5	21.4
6	14.8	16.1	15.3	16.6	17.0
7	15.4	21.2	22.2	25.8	21.0
8	16.9	21.5	22.1	25.3	24.1
9	23.3	22.6	23.8	27.5	29.0
10	24.4	22.3	25.2	25.0	24.9
Mean	**20.5**	**21.5**	**22.1**	**23.7**	**24.2**
SD	**4.5**	**3.3**	**3.2**	**3.6**	**4.6**

Appendix 4.2.6 Peak of mediolateral shear force (% BW)

Left stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	7.0	6.6	9.7	9.7	10.0
2	10.2	9.9	9.3	8.0	11.2
3	7.9	8.5	7.4	8.2	8.5
4	7.8	8.2	7.7	9.4	8.9
5	5.5	5.3	6.3	6.6	5.5
6	7.3	7.5	8.6	8.4	9.4
7	7.4	8.2	8.2	10.5	9.3
8	4.6	6.9	6.1	6.8	6.1
9	7.5	6.1	6.4	6.8	6.4
10	7.6	8.1	8.5	7.8	10.4
Mean	**7.3**	**7.5**	**7.8**	**8.2**	**8.6**
SD	**1.5**	**1.3**	**1.3**	**1.3**	**1.9**

Right stride	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	5.3	6.2	10.5	7.7	9.1
2	6.1	5.7	5.4	6.2	6.2
3	7.8	9.8	9.2	9.1	9.2
4	7.6	7.1	8.7	9.4	9.1
5	10.0	8.0	7.7	10.0	10.7
6	7.5	7.1	5.7	7.1	7.0
7	6.9	8.5	9.5	8.6	9.7
8	6.9	7.9	7.4	7.6	8.2
9	7.2	6.8	7.2	7.9	6.7
10	11.0	9.7	10.6	11.6	11.1
Mean	**7.6**	**7.7**	**8.2**	**8.5**	**8.7**
SD	**1.7**	**1.4**	**1.8**	**1.6**	**1.7**

Appendix 4.3
Experimental data of peak EMG amplitudes under backpack carriage during walking

Appendix 4.3.1 Rectus abdominis (% EMG_{MVC})

Left	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	2.1	2.3	2.7	3.5	3.0
2	2.2	2.8	3.0	2.1	4.4
3	2.8	2.1	2.6	5.0	7.7
4	5.2	4.5	7.9	12.4	10.1
5	3.7	4.4	5.4	7.6	4.5
6	1.7	2.6	4.4	3.6	13.5
7	5.1	5.1	4.7	6.9	7.2
8	2.0	2.2	2.4	3.7	4.9
9	3.4	3.3	4.1	4.8	7.3
10	2.0	1.3	3.5	4.3	5.1
Mean	**3.0**	**3.1**	**4.1**	**5.4**	**6.8**
SD	**1.3**	**1.2**	**1.7**	**3.0**	**3.1**

Right	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	2.5	3.2	4.1	4.8	3.6
2	2.7	2.1	2.3	2.8	4.8
3	1.8	1.6	1.7	2.7	3.6
4	7.9	9.2	12.2	17.7	12.3
5	4.5	5.6	6.7	7.9	7.2
6	2.0	1.2	3.8	6.8	11.6
7	4.9	5.9	7.2	9.9	10.1
8	2.5	3.8	3.3	4.9	8.7
9	1.9	2.3	2.2	2.5	3.4
10	2.7	3.5	3.9	7.0	6.4
Mean	**3.3**	**3.8**	**4.7**	**6.7**	**7.2**
SD	**1.9**	**2.4**	**3.2**	**4.6**	**3.4**

Appendix 4.3.2 External oblique (% EMG_{MVC})

Left	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	9.5	9.7	9.1	12.7	11.5
2	3.2	2.8	4.3	5.5	4.2
3	4.1	2.8	4.7	5.7	4.7
4	13.1	19.3	19.4	23.2	18.3
5	5.8	5.8	7.9	10.9	6.3
6	13.7	12.6	13.1	19.0	14.3
7	6.5	6.2	6.3	6.7	4.3
8	1.7	1.8	2.6	3.8	3.7
9	7.8	9.1	8.8	10.3	9.0
10	1.6	2.4	2.0	3.3	5.3
Mean	**6.7**	**7.3**	**7.8**	**10.1**	**8.2**
SD	**4.3**	**5.6**	**5.3**	**6.6**	**5.0**

Right	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	6.9	6.3	7.5	9.4	5.6
2	2.2	2.0	2.4	2.9	3.4
3	4.0	2.8	4.8	5.3	8.2
4	13.4	16.1	21.2	23.9	18.5
5	4.9	4.1	5.6	7.2	6.9
6	10.7	14.1	14.8	17.5	15.3
7	5.3	5.0	8.3	8.2	8.0
8	2.5	4.3	4.5	6.9	6.3
9	10.4	10.4	11.5	14.8	14.0
10	2.6	3.8	3.9	4.8	6.5
Mean	**6.3**	**6.9**	**8.5**	**10.1**	**9.3**
SD	**3.9**	**4.9**	**5.8**	**6.6**	**4.9**

Appendix 4.3.3 Internal oblique (% EMG_{MVC})

Left	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	5.9	3.9	3.1	6.2	3.8
2	6.0	8.1	8.1	5.9	19.3
3	8.5	8.4	18.0	13.0	15.4
4	7.0	6.1	8.6	10.0	7.9
5	4.3	4.9	4.5	4.7	6.2
6	11.0	17.7	12.8	17.2	14.2
7	3.5	5.2	6.6	11.7	10.0
8	12.4	14.0	12.1	12.7	9.3
9	14.8	14.9	13.3	15.6	16.3
10	4.2	6.0	7.8	7.9	6.0
Mean	**7.8**	**8.9**	**9.5**	**10.5**	**10.8**
SD	**3.8**	**4.8**	**4.5**	**4.3**	**5.2**

Right	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	6.8	3.7	3.2	8.3	4.8
2	5.6	5.8	6.5	5.4	12.4
3	13.0	8.2	17.2	18.2	15.8
4	4.7	5.4	7.5	8.7	6.9
5	4.4	4.4	6.0	7.9	6.3
6	13.6	17.9	18.1	15.8	23.0
7	4.0	3.5	6.8	13.7	9.1
8	8.4	7.4	6.1	8.9	10.0
9	15.1	16.0	15.1	15.5	16.0
10	14.8	13.1	10.9	17.8	10.2
Mean	**9.0**	**8.5**	**9.7**	**12.0**	**11.5**
SD	**4.6**	**5.3**	**5.3**	**4.7**	**5.5**

Appendix 4.3.4 Latissimus dorsi (% EMG_{MVC})

Left	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	3.8	3.3	8.5	3.0	2.2
2	7.2	6.3	10.0	7.2	9.2
3	4.3	3.8	3.4	4.7	6.8
4	3.1	2.5	2.6	3.5	3.2
5	3.7	2.4	2.6	2.4	5.6
6	2.5	3.3	3.6	3.4	4.1
7	4.3	1.1	2.6	2.9	3.7
8	7.0	6.0	7.6	4.5	4.2
9	1.7	2.1	1.8	2.1	3.2
10	7.6	9.7	9.4	7.9	8.1
Mean	**4.5**	**4.1**	**5.2**	**4.2**	**5.0**
SD	**2.1**	**2.6**	**3.2**	**2.0**	**2.3**

Right	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	5.0	4.4	4.1	2.9	2.0
2	6.7	5.3	9.0	6.6	3.9
3	1.3	1.7	1.6	1.6	2.1
4	4.0	2.7	2.8	3.6	4.5
5	3.0	2.8	2.2	2.6	4.8
6	4.5	6.9	4.9	3.5	3.0
7	2.3	1.9	1.7	1.3	1.7
8	1.4	1.2	1.0	1.3	1.3
9	4.2	3.6	2.8	2.9	2.5
10	9.0	6.0	9.2	6.0	8.1
Mean	**4.1**	**3.7**	**3.9**	**3.2**	**3.4**
SD	**2.4**	**1.9**	**3.0**	**1.8**	**2.0**

Appendix 4.3.5 Thoracic erector spinae (% EMG_{MVC})

Left	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	7.0	5.0	3.1	2.6	2.1
2	12.0	9.9	14.5	12.1	8.1
3	4.3	4.8	3.5	5.1	8.5
4	7.0	5.5	5.9	8.0	6.5
5	4.8	3.0	4.1	6.6	7.8
6	10.2	9.8	6.8	5.8	4.9
7	4.8	5.2	6.1	5.3	7.9
8	11.6	11.6	9.0	8.9	7.3
9	2.1	1.9	1.5	2.0	1.7
10	5.4	8.6	10.1	9.7	11.1
Mean	**6.9**	**6.5**	**6.5**	**6.6**	**6.6**
SD	**3.3**	**3.2**	**3.9**	**3.1**	**2.9**

Right	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	6.5	6.7	5.1	3.5	4.3
2	11.0	10.5	14.0	11.0	8.4
3	6.2	4.8	5.3	5.0	8.9
4	5.2	4.3	7.4	3.6	5.1
5	7.0	5.9	5.1	4.6	7.9
6	9.4	10.2	4.8	5.8	6.9
7	7.1	6.1	4.8	4.6	7.8
8	5.2	5.0	16.2	6.3	10.1
9	8.2	9.0	6.8	7.3	6.7
10	5.7	4.7	4.6	11.5	6.3
Mean	**7.2**	**6.7**	**7.4**	**6.3**	**7.2**
SD	**1.9**	**2.3**	**4.2**	**2.8**	**1.7**

Appendix 4.3.6 Lumbar erector spinae (% EMG_{MVC})

Left	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	7.5	7.4	4.4	4.5	2.9
2	7.7	7.0	7.4	5.9	7.0
3	8.7	7.5	6.4	7.5	10.3
4	6.8	5.2	2.8	2.5	6.8
5	8.3	7.4	5.6	7.0	4.8
6	13.8	11.7	8.6	10.0	9.5
7	7.2	8.3	5.6	4.8	5.7
8	9.1	5.3	6.3	4.8	4.1
9	6.0	8.4	8.3	8.9	8.5
10	12.1	14.6	12.2	11.8	10.8
Mean	**8.7**	**8.3**	**6.8**	**6.8**	**7.0**
SD	**2.4**	**2.9**	**2.6**	**2.8**	**2.7**

Right	Load (% BW)				
Participant	**0**	**5**	**10**	**15**	**20**
1	10.9	10.5	10.9	8.5	9.9
2	9.6	8.8	5.8	8.2	8.1
3	8.5	7.5	6.8	7.5	9.8
4	6.5	4.2	4.1	4.2	4.4
5	7.9	6.8	7.4	7.0	6.1
6	12.1	14.5	11.1	10.6	9.5
7	8.1	8.0	7.9	3.9	5.8
8	9.0	7.3	9.3	5.5	9.9
9	6.2	5.7	3.6	4.1	5.8
10	5.5	5.0	6.4	6.7	7.7
Mean	**8.4**	**7.8**	**7.3**	**6.6**	**7.7**
SD	**2.1**	**3.0**	**2.6**	**2.2**	**2.1**

Appendix 4.4
Experimental data of peak lumbosacral joint forces under backpack carriage during walking

Appendix 4.4.1 Compression (% BW)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	197.2	190.4	216.9	241.1	270.5
2	171.5	174.4	178.5	197.5	208.5
3	205.9	191.7	210.5	252.3	274.2
4	180.9	188.4	214.6	207.9	249.8
5	193.9	178.8	181.4	246.0	273.4
6	190.0	184.2	229.0	253.7	265.2
7	190.1	189.0	213.9	223.1	242.0
8	204.6	168.8	221.4	249.4	270.3
9	188.2	211.6	208.7	231.2	253.8
10	231.5	218.3	222.4	239.8	242.4
Mean	**195.4**	**189.6**	**209.7**	**234.2**	**255.0**
SD	**16.3**	**15.4**	**16.8**	**19.2**	**20.6**

Appendix 4.4.2 Anterior shear (% BW)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	27.5	24.2	35.8	29.9	28.8
2	31.1	33.1	29.5	32.4	45.3
3	33.6	37.5	42.2	39.4	48.8
4	15.3	11.7	12.3	13.2	24.5
5	21.4	26.2	25.1	25.9	30.2
6	24.0	23.3	23.7	37.0	31.8
7	23.9	21.9	16.9	25.0	25.3
8	22.6	22.6	24.9	27.6	34.1
9	28.8	24.8	25.2	22.4	26.1
10	21.9	31.5	29.7	30.4	39.3
Mean	**25.0**	**25.7**	**26.5**	**28.3**	**33.4**
SD	**5.3**	**7.1**	**8.6**	**7.5**	**8.5**

Appendix 4.4.3 Posterior shear (% BW)

Participant	Load (% BW)				
	0	5	10	15	20
1	10.3	11.1	8.5	9.4	11.3
2	23.7	23.7	24.7	28.0	32.9
3	19.5	25.7	29.5	40.1	38.7
4	23.0	27.6	31.2	29.0	20.0
5	30.8	32.2	39.9	42.3	39.9
6	11.2	13.2	20.7	20.3	23.2
7	10.5	12.2	17.7	12.3	26.0
8	19.6	16.3	21.1	17.3	23.4
9	19.1	22.8	25.3	25.6	22.9
10	28.5	24.0	26.6	26.0	26.1
Mean	**19.6**	**20.9**	**24.5**	**25.0**	**26.4**
SD	**7.3**	**7.2**	**8.4**	**10.7**	**8.7**

Appendix 4.4.4 Mediolateral shear (% BW)

Participant	Load (% BW)				
	0	**5**	**10**	**15**	**20**
1	16.0	12.0	23.0	24.7	27.5
2	12.7	16.2	14.9	17.9	25.7
3	17.1	19.9	21.1	20.3	23.5
4	16.9	14.8	18.4	17.9	14.8
5	12.2	16.0	22.4	24.2	25.0
6	16.0	15.2	16.9	15.9	17.5
7	16.7	16.7	18.9	18.6	19.1
8	9.5	11.9	12.1	10.7	14.7
9	16.5	18.5	20.1	21.6	19.5
10	17.6	13.8	10.3	14.4	13.1
Mean	**15.1**	**15.5**	**17.8**	**18.6**	**20.0**
SD	**2.7**	**2.6**	**4.3**	**4.3**	**5.1**

Appendix 5.1
Experimental data of Peak Compression Force Deviation from EMGAO Approach (% BW)
(3 trials x 5 backpack loading conditions x 10 participants = 150 data points)

Data	REMGAO	DLOPT	SLOPT	Data	REMGAO	DLOPT	SLOPT
1	4.8	-7.4	-19.5	**26**	-1.0	5.1	-20.2
2	-2.6	-13.0	-23.1	**27**	-7.8	9.7	-16.0
3	1.2	-6.7	-18.4	**28**	-5.8	-6.6	-17.7
4	6.0	0.1	-16.5	**29**	-6.2	0.4	-14.8
5	-2.5	-12.6	-27.2	**30**	-9.8	1.9	-16.7
6	1.9	-9.9	-23.6	**31**	-9.8	-30.3	-39.5
7	10.0	-0.6	-21.9	**32**	1.9	-0.7	-12.8
8	2.5	5.0	-17.1	**33**	5.6	-2.4	-12.6
9	1.6	-7.7	-19.0	**34**	0.3	-8.2	-19.7
10	2.1	-7.5	-23.7	**35**	1.1	-5.2	-17.8
11	-2.8	-7.5	-26.1	**36**	2.3	-5.1	-17.1
12	1.8	-8.1	-24.2	**37**	-4.1	-6.7	-17.5
13	1.5	-1.6	-21.1	**38**	-0.8	-29.7	-41.7
14	-0.2	-3.0	-21.4	**39**	-0.4	-7.3	-20.0
15	0.3	-3.1	-20.9	**40**	0.7	1.0	-13.4
16	5.9	9.5	-16.5	**41**	2.0	4.8	-10.1
17	-9.6	-8.3	-26.5	**42**	0.7	-7.2	-15.5
18	2.7	-9.0	-13.4	**43**	0.9	-0.8	-12.5
19	-0.2	2.8	-13.9	**44**	1.6	-20.1	-31.3
20	6.6	-2.7	-19.2	**45**	8.4	-8.2	-26.9
21	-5.6	-1.4	-14.5	**46**	-2.1	-3.9	-20.7
22	4.4	4.7	-17.5	**47**	1.4	-3.4	-19.7
23	2.8	0.8	-20.1	**48**	0.9	-1.2	-18.5
24	3.0	1.9	-20.1	**49**	-1.6	1.1	-19.1
25	3.2	7.9	-14.6	**50**	-4.5	-11.1	-22.2

Appendix 5.1 (cont'd)
Experimental data of Peak Compression Force Deviation from EMGAO Approach (% BW)

Data	REMGAO	DLOPT	SLOPT	Data	REMGAO	DLOPT	SLOPT
51	-1.8	0.2	-19.3	**76**	0.1	-8.5	-18.6
52	-1.7	-14.9	-29.1	**77**	3.3	-13.6	-23.6
53	-0.8	-12.7	-26.9	**78**	0.3	-8.6	-18.3
54	-0.5	-13.0	-27.1	**79**	0.2	-19.8	-25.9
55	-5.9	-11.1	-19.9	**80**	2.1	-12.9	-19.1
56	-0.9	3.9	-16.3	**81**	2.5	-12.4	-19.3
57	-7.9	-13.0	-21.9	**82**	-1.0	-23.5	-31.8
58	-1.3	-3.2	-20.3	**83**	-1.2	-12.3	-29.9
59	-3.6	-8.6	-19.8	**84**	-6.9	-23.1	-36.8
60	-0.5	1.3	-16.3	**85**	-1.7	-12.3	-28.7
61	0.7	-8.3	-19.9	**86**	2.0	-9.8	-27.4
62	1.0	-4.3	-15.0	**87**	2.0	-9.4	-27.8
63	1.5	-9.7	-19.3	**88**	-4.5	-17.3	-33.2
64	1.5	-7.0	-14.3	**89**	-0.5	-9.3	-26.2
65	1.1	-6.1	-14.8	**90**	-0.5	-7.3	-25.0
66	0.9	-9.1	-14.5	**91**	6.2	-1.2	-19.6
67	-1.2	11.7	-16.0	**92**	10.8	-2.7	-24.4
68	-0.9	4.0	-16.5	**93**	1.6	0.6	-13.7
69	0.6	17.7	-10.2	**94**	0.7	-5.7	-12.0
70	0.3	-7.0	-16.8	**95**	-2.1	0.2	-18.1
71	0.7	-6.9	-17.0	**96**	0.3	-8.5	-14.7
72	0.6	-6.0	-16.7	**97**	4.8	-6.7	-16.8
73	2.4	-1.9	-11.2	**98**	1.2	-14.0	-26.0
74	3.1	-2.3	-12.0	**99**	2.2	-11.2	-22.8
75	2.5	-2.0	-11.4	**100**	-1.4	-7.0	-23.7

Appendix 5.1 (cont'd)

Experimental data of Peak Compression Force Deviation from EMGAO Approach (% BW)

Data	REMGAO	DLOPT	SLOPT	Data	REMGAO	DLOPT	SLOPT
101	-1.6	-7.7	-25.3	**126**	-4.3	-8.6	-19.9
102	0.6	-11.5	-24.2	**127**	-5.9	-5.1	-19.8
103	-9.4	-16.2	-25.1	**128**	-7.1	-10.9	-21.5
104	0.4	1.9	-13.4	**129**	2.7	-16.1	-19.8
105	0.4	2.7	-11.1	**130**	3.5	-5.6	-18.3
106	1.1	6.7	-5.9	**131**	-6.0	-10.3	-21.3
107	1.1	4.5	-9.2	**132**	0.3	-4.1	-15.6
108	-1.1	0.9	-9.2	**133**	-1.3	-7.3	-17.9
109	3.6	-12.5	-28.3	**134**	-0.3	-0.3	-20.2
110	6.1	-7.3	-20.6	**135**	0.7	-13.3	-26.5
111	-10.6	-13.5	-21.1	**136**	4.5	9.6	-10.1
112	-8.0	-10.7	-24.3	**137**	-2.4	18.1	-7.3
113	3.7	2.3	-11.8	**138**	-11.3	11.8	-7.0
114	7.3	-12.6	-27.5	**139**	-1.9	-6.0	-9.2
115	0.3	4.7	-12.2	**140**	1.8	5.3	-6.8
116	0.1	0.9	-12.8	**141**	11.8	2.3	-16.5
117	-0.5	-4.0	-13.3	**142**	0.0	-1.4	-12.7
118	3.7	-5.4	-17.8	**143**	6.1	8.5	-14.3
119	-2.9	1.6	-11.1	**144**	0.2	8.4	-9.7
120	-0.8	-0.3	-12.5	**145**	-2.9	6.9	-6.3
121	1.3	-22.5	-27.0	**146**	4.9	5.1	-15.3
122	0.5	-14.3	-18.4	**147**	3.7	-6.5	-20.7
123	0.6	-21.0	-25.7	**148**	-10.6	6.2	-20.9
124	-8.3	-16.7	-25.1	**149**	-6.2	8.4	-25.3
125	-2.5	-22.3	-27.0	**150**	2.0	-11.3	-23.5

Appendix 6.1
Experimental data of scoliotic spinal curvatures under single-strap cross-chest bag carriage in upright stance

Appendix 6.1.1 Participants 1-6

Loaded on the ipsilateral side of apex location	No-load	Asymmetric Load (% BW)				
	0	**2.5**	**5**	**7.5**	**10**	**12.5**
Participant 1	22.2	22.5	24.3	17.5	23.3	26.5
Participant 2	21.9	21.5	25.3	19.5	23.3	27.0
Participant 3	17.2	18.8	13.8	23.0	14.3	15.5
Participant 4	14.3	16.5	16.2	15.1	14.4	11.4
Participant 5	15.9	26.8	20.8	15.5	19.1	18.3
Participant 6	12.6	6.4	7.2	10.8	16.6	21.4
Mean	**17.4**	**18.8**	**17.9**	**16.9**	**18.5**	**20.0**
SD	**4.0**	**7.0**	**6.9**	**4.1**	**4.1**	**6.2**
Loaded on the contralateral side of apex location	**No-load**	**Asymmetric Load (% BW)**				
	0	**2.5**	**5**	**7.5**	**10**	**12.5**
Participant 1	22.2	21.2	21.6	19.8	17.4	16.8
Participant 2	21.9	21.4	19.2	19.3	17.1	16.5
Participant 3	17.2	14.7	11.0	13.7	14.7	16.4
Participant 4	14.3	10.8	12.6	11.3	13.7	13.1
Participant 5	15.9	14.2	15.4	14.5	15.7	15.7
Participant 6	12.6	5.9	6.2	7.3	5.3	4.0
Mean	**17.4**	**14.7**	**14.3**	**14.3**	**14.0**	**13.8**
SD	**4.0**	**6.0**	**5.6**	**4.8**	**4.5**	**5.0**

Appendix 6.1.2 Participant 7

Loaded on the contralateral side of apex location	Asymmetric Load (% BW)											
	Minor Lumbar Curve						Major Thoracic Curve					
	0	2.5	5	7.5	10	12.5	0	2.5	5	7.5	10	12.5
Trial 1	8.9	7.6	9.6	8.6	6.2	6.5	17.5	12.5	9.2	5.8	8.5	6.5
Trial 2	6.3	7.9	7.9	9.4	7.3	8.8	16.6	10.0	7.1	6.2	9.3	7.9
Trial 3	7.5	9.1	8.0	7.2	6.9	9.6	18.2	8.6	8.0	7.7	7.6	9.9
Mean	**7.6**	**8.2**	**8.5**	**8.4**	**6.8**	**8.3**	**17.4**	**10.4**	**8.1**	**6.6**	**8.5**	**8.1**
SD	**1.3**	**0.8**	**1.0**	**1.1**	**0.6**	**1.6**	**0.8**	**2.0**	**1.1**	**1.0**	**0.9**	**1.7**

References

Abe, D., Yanagawa, K., & Niihata, S. (2004). Effects of load carriage, load position, and walking speed on energy cost of walking. Applied ergonomics, 35(4), 329-335.

Adeyemi, A. J., Rohani, J. M., & Rani, M. R. A. (2017). Backpack-back pain complexity and the need for multifactorial safe weight recommendation. Applied ergonomics, 58, 573-582.

Al-Hazzaa, H. M. (2006). School backpack. How much load do Saudi school boys carry on their shoulders?. Saudi medical journal, 27(10), 1567-1571.

Al-Khabbaz, Y. S., Shimada, T., & Hasegawa, M. (2008). The effect of backpack heaviness on trunk-lower extremity muscle activities and trunk posture. Gait & posture, 28(2), 297-302.

Amarantini, D., Rao, G., & Berton, E. (2010). A two-step EMG-and-optimization process to estimate muscle force during dynamic movement. Journal of Biomechanics, 43(9), 1827-1830.

Anders, C., Brose, G., Hofmann, G. O., & Scholle, H. C. (2008). Evaluation of the EMG–force relationship of trunk muscles during whole body tilt. Journal of Biomechanics, 41(2), 333-339.

Aouni, B., & Kettani, O. (2001). Goal programming model: A glorious history and a promising future. European Journal of Operational Research, 133(2), 225-231.

Arenas, M., Bilbao, A., Caballero, R., Gómez, T., Rodriguez, M. V., & Ruiz, F. (2002). Analysis via goal programming of the minimum achievable stay in surgical waiting lists. Journal of the Operational Research society, 53(4), 387-396.

Arjmand, N., Gagnon, D., Plamondon, A., Shirazi-Adl, A., & Lariviere, C. (2009). Comparison of trunk muscle forces and spinal loads estimated by two biomechanical models. Clinical Biomechanics, 24(7), 533-541.

Arkin, A. M., & Katz, J. F. (1956). The effects of pressure on epiphyseal growth. J Bone Joint Surg Am, 38(5), 1056-1076.

Aroeira, R. M. C., Leal, J. S., & de Melo Pertence, Antnio Eustquio. (2011). New method of scoliosis assessment: Preliminary results using computerized photogrammetry. Spine, 36(19), 1584-1591.

Atreya, S., Singh, U., Ribeiro, R., Anand, S., Jaryal, A., & Srivastava, S. (2010, December). Effect of load distribution of the backpack-a design study. In Systems in Medicine and Biology (ICSMB), 2010 International Conference on (pp. 183-186). IEEE.

Attwells, R. L., Birrell, S. A., Hooper, R. H., & Mansfield, N. J. (2006). Influence of carrying heavy loads on soldiers' posture, movements and gait. Ergonomics, 49(14), 1527-1537.

Ayyappa, E. (1997). Normal Human Locomotion, Part 1: Basic Concepts and Terminology. JPO: Journal of Prosthetics and Orthotics, 9(1), 10-17.

Bae, T. S., & Mun, M. (2010). Effect of lumbar lordotic angle on lumbosacral joint during isokinetic exercise: A simulation study. Clinical Biomechanics, 25(7), 628-635.

Balasubramanian, S. (2013). Comparison of Angle Measurements between Vicon and Myomotion System. http://www.noraxon.com/ wp-conntent/ uploads/

2015/ 06/ myoMOTION-Validation-Study.pdf.

Bampouras, T. M., & Dewhurst, S. (2016). Carrying shopping bags does not alter static postural stability and gait parameters in healthy older females. Gait & posture, 46, 81-85.

Barkhordari, A., Ehrampoush, M. H., Barkhordari, M., Derakhshi, F., Barkhoradri, M., & Mirzaii, M. (2013). Assessment of school backpack weight and other characteristics in elementary schools, Yazd, Iran. J Health Res, 2, 2-7.

Bastien, G. J., Willems, P. A., Schepens, B., & Heglund, N. C. (2005). Effect of load and speed on the energetic cost of human walking. European journal of applied physiology, 94(1-2), 76-83.

Bastien, G. J., Willems, P. A., Schepens, B., & Heglund, N. C. (2016). The mechanics of head-supported load carriage by Nepalese porters. Journal of Experimental Biology, 219(22), 3626-3634.

Bean, J. C., Chaffin, D. B., & Schultz, A. B. (1988). Biomechanical model calculation of muscle contraction forces: a double linear programming method. Journal of Biomechanics, 21(1), 59-66.

Bedale, M. (1924). Comparison of the energy expenditures of a woman carrying loads in eight different positions. Industrial Hygiene Research Board Report No. 29. Medical Research Council, London.

Birrell, S. A., & Haslam, R. A. (2009). The effect of military load carriage on 3-D lower limb kinematics and spatiotemporal parameters. Ergonomics, 52(10), 1298-1304.

Birrell, S. A., Hooper, R. H., & Haslam, R. A. (2007). The effect of military load carriage on ground reaction forces. Gait & posture, 26(4), 611-614.

Bloswick, D. (1999). Climbing biomechanics. Biomechanics in ergonomics, 335-349.

Bottoms, K. E., & Bartlett, E. T. (1975). Resource allocation through goal programming. Journal of range management, 442-447.

Boyer, J., Amin, N., Taddonio, R., & Dozor, A. J. (1996). Evidence of airway obstruction in children with idiopathic scoliosis. Chest, 109(6), 1532-1535.

Brackley, H. M., & Stevenson, J. M. (2004). Are children's backpack weight limits enough?: A critical review of the relevant literature. Spine, 29(19), 2184-2190.

Brackley, H. M., Stevenson, J. M., & Selinger, J. C. (2009). Effect of backpack load placement on posture and spinal curvature in prepubescent children. Work, 32(3), 351-360.

Bruno, A. G., Anderson, D. E., D'Agostino, J., & Bouxsein, M. L. (2012). The effect of thoracic kyphosis and sagittal plane alignment on vertebral compressive loading. Journal of Bone and Mineral Research, 27(10), 2144-2151.

Buchanan, T. S., & Shreeve, D. A. (1996). An Evaluation of Optimization Techniques for the prediction of muscle activation patterns during isometric tasks. Journal of Biomechanical Engineering, 118, 565-574.

Brzęk, A., Dworrak, T., Strauss, M., Sanchis-Gomar, F., Sabbah, I., Dworrak, B., & Leischik, R. (2017). The weight of pupils' schoolbags in early school age and its influence on body posture. BMC musculoskeletal disorders, 18(1), 117.

Byl, N. N., Holland, S., Jurek, A., & Hu, S. S. (1997). Postural imbalance and vibratory sensitivity in patients with idiopathic scoliosis: Implications for treatment.

Journal of Orthopaedic & Sports Physical Therapy, 26(2), 60-68.

Callaghan, J. P., Patla, A. E., & McGill, S. M. (1999). Low back three-dimensional joint forces, kinematics, and kinetics during walking. Clinical Biomechanics, 14(3), 203-216.

Cappellini, G., Ivanenko, Y. P., Poppele, R. E., & Lacquaniti, F. (2006). Motor patterns in human walking and running. Journal of neurophysiology, 95(6), 3426-3437.

Carbone, P. D., Carlton, S. D., Stierli, M., & Orr, R. M. (2014). The impact of load carriage on the marksmanship of the tactical police officer: a pilot study. Journal of Australian Strength and Conditioning, 22(2), 50-57.

Carre, L.T. (1908). Historical review of the load of the foot soldiers. La Revue D'Infanterie, 1907. Translated by P. H. Miles. Infantry Joural, 5. 757-916. 1907. Reprinted by the Office of the Quartermaster General Research and Development Branch, Textile, Clothing and Footwear Division. Tentage and Equipage Series. Report Number 8.

Carrier, J., Aubin, C., Villemure, I., & Labelle, H. (2004). Biomechanical modelling of growth modulation following rib shortening or lengthening in adolescent idiopathic scoliosis. Medical and Biological Engineering and Computing, 42(4), 541-548.

Carrizosa, E., & Fliege, J. (2002). Generalized goal programming: Polynomial methods and applications. Mathematical Programming, 93(2), 281-303.

Chansirinukor, W., Wilson, D., Grimmer, K., & Dansie, B. (2001). Effects of backpacks on students: measurement of cervical and shoulder posture. Australian Journal of physiotherapy, 47(2), 110-116.

Charnes, A., & Cooper, W. W. (1961). Management models and industrial applications of linear programming (Vol. 1, p. 2). New York: Wiley.

Charnes, A., Cooper, W. W., & Ferguson, R. O. (1955). Optimal estimation of executive compensation by linear programming. Management science, 1(2), 138-151.

Cheng, C. K., Chen, H. H., Chen, C. S., & Lee, S. J. (1998). Influences of walking speed change on the lumbosacral joint force distribution. Bio-medical materials and engineering, 8(3, 4), 155-165.

Chiang, H. Y., Jacobs, K., & Orsmond, G. (2006). Gender-age environmental associates of middle school students' low back pain. Work, 26(2), 197-206.

Cholewicki, J., & McGill, S. M. (1994). EMG assisted optimization: a hybrid approach for estimating muscle forces in an indeterminate biomechanical model. Journal of Biomechanics, 27(10), 1287-1289.

Cholewicki, J., McGill, S. M., & Norman, R. W. (1995). Comparison of muscle forces and joint load from an optimization and EMG assisted lumbar spine model: towards development of a hybrid approach. Journal of Biomechanics, 28(1), 321-331.

Chow, D. H. K., Hin, C. K. F., Ou, D., & Lai, A. (2011b). Carry-over effects of backpack carriage on trunk posture and repositioning ability. International Journal of Industrial Ergonomics, 41(5), 530-535.

Chow, D. H. K., Kwok, M. L., Au-Yang, A. C., Holmes, A. D., Cheng, J. C., Yao, F. Y., & Wong, M. S. (2005a). The effect of backpack load on the gait of normal adolescent girls. Ergonomics, 48(6), 642-656.

Chow, D. H. K., Kwok, M. L., Cheng, J. C., Lao, M. L., Holmes, A. D., Au-Yang, A., Wong, M. S. (2006a). The effect of backpack weight on the standing posture and balance of schoolgirls with adolescent idiopathic scoliosis and normal controls. Gait & Posture, 24(2), 173-181.

Chow, D. H. K., Kwok, M. L., Au-Yang, A. C., Holmes, A. D., Cheng, J. C., Yao, F. Y., & Wong, M. S. (2006b). The effect of load carriage on the gait of girls with adolescent idiopathic scoliosis and normal controls. Medical engineering & physics, 28(5), 430-437.

Chow, D. H. K., Leung, K. T. Y., & Holmes, A. D. (2007). Changes in spinal curvature and proprioception of schoolboys carrying different weights of backpack. Ergonomics, 50(12), 2148-2156.

Chow, D. H. K., Li, M. F., Lai, A., & Pope, M. H. (2011a). Effect of load carriage on spinal compression. International Journal of Industrial Ergonomics, 41(3), 219-223.

Chow, D. H. K., Ng, X. H., Holmes, A. D., Cheng, J. C., Yao, F. Y., & Wong, M. S. (2005b). Effects of backpack loading on the pulmonary capacities of normal schoolgirls and those with adolescent idiopathic scoliosis. Spine, 30(21), E654.

Chow, D. H. K., Ou, Z. Y., Wang, X. G., & Lai, A. (2010). Short-term effects of backpack load placement on spine deformation and repositioning error in schoolchildren. Ergonomics, 53(1), 56-64.

Chow, D. H. K., Ting, J. M. L., Pope, M. H., & Lai, A. (2009). Effects of backpack load placement on pulmonary capacities of normal schoolchildren during upright stance. International journal of industrial ergonomics, 39(5), 703-707.

Chow, D. H. K., Wang, C., & Pope, M. H. (2014). Effects of backpack carriage on lumbopelvic control: A dynamical systems analysis. International Journal of Industrial Ergonomics, 44(4), 493-498.

Conolly, M., Elder, C., & Dawes, J. (2015). Needs Analysis for Mountain Search and Rescue. Strength & Conditioning Journal, 37(4), 35-42.

Cook, C. (2007). Orthopedic manual therapy: an evidence-based approach. Upper Saddle River, N.J. Pearson/Prentice Hall.

Corrigan, L. P., & Li, J. X. (2014). The effect of unilateral hockey bag carriage on the muscle activities of the trunk and lower limb of young healthy males during gait. Research in Sports Medicine, 22(1), 23-35.

Costello, K. E., Matrangola, S. L., & Madigan, M. L. (2012). Independent effects of adding weight and inertia on balance during quiet standing. Biomedical engineering online, 11(1), 20.

Crowninshield, R. D., & Brand, R. A. (1981). A physiologically based criterion of muscle force prediction in locomotion. Journal of Biomechanics, 14(11), 793-801.

Dantzig, G. B. (1949). Programming of interdependent activities: II mathematical model. Econometrica, Journal of the Econometric Society, 200-211.

Dames, K. D., & Smith, J. D. (2015). Effects of load carriage and footwear on spatiotemporal parameters, kinematics, and metabolic cost of walking. Gait & posture, 42(2), 122-126.

Deckro, R. F., & Hebert, J. E. (1988). Polynomial goal programming: A procedure for modeling preference trade-offs. Journal of Operations Management, 7(3-4), 149-164.

Delice, E. K., & Güngör, Z. (2011). A mixed integer goal programming model for discrete values of design requirements in QFD. International journal of production research, 49(10), 2941-2957.

Den Boer, W. A., Anderson, P. G., v. Limbeek, J., & Kooijman, M. (1999). Treatment of idiopathic scoliosis with side-shift therapy: An initial comparison with a brace treatment historical cohort. European Spine Journal, 8(5), 406-410.

DeTroy, C., Jonkers, I., De Becker, A., Lenaerts, G., & Spaepen, A. (2007). Evaluation of the effect of backpack load and position during standing and walking using biomechanical, physiological and subjective measures. Ergonomics, 50(5), 728-742.

Devroey, C., Jonkers, I., De Becker, A., Lenaerts, G., & Spaepen, A. (2007). Evaluation of the effect of backpack load and position during standing and walking using biomechanical, physiological and subjective measures. Ergonomics, 50(5), 728-742.

Dianat, I., Javadivala, Z., Asghari-Jafarabadi, M., Asl Hashemi, A., & Haslegrave, C. M. (2013). The use of schoolbags and musculoskeletal symptoms among primary school children: are the recommended weight limits adequate?. Ergonomics, 56(1), 79-89.

Dockrell, S., Simms, C., & Blake, C. (2013). Schoolbag weight limit: can it be defined?. Journal of School Health, 83(5), 368-377.

Dominelli, P. B., Sheel, A. W., & Foster, G. E. (2012). Effect of carrying a weighted backpack on lung mechanics during treadmill walking in healthy men. European journal of applied physiology, 112(6), 2001-2012.

Dreischarf, M., Shirazi-Adl, A., Arjmand, N., Rohlmann, A., & Schmidt, H. (2016). Estimation of loads on human lumbar spine: a review of in vivo and computational model studies. Journal of Biomechanics, 49(6), 833-845.

Drzał-Grabiec, J., Snela, S., Rachwał, M., Podgórska, J., & Rykała, J. (2015). Effects of carrying a backpack in an asymmetrical manner on the asymmetries of the trunk and parameters defining lateral flexion of the spine. Human factors, 57(2), 218-226.

Dul, J., Johnson, G. E., Shiavi, R., & Townsend, M. A. (1984). Muscular synergism—II. A minimum-fatigue criterion for load sharing between synergistic muscles. Journal of Biomechanics, 17(9), 675-684.

Durbach, I. N., & Stewart, T. J. (2012). Modeling uncertainty in multi-criteria decision analysis. European Journal of Operational Research, 223(1), 1-14.

Durmala, J., Dobosiewicz, K., Kotwicki, T., & Jendrzejek, H. (2003). Influence of asymmetric mobilisation of the trunk on the cobb angle and rotation in idiopathic scoliosis in children and adolescents. Ortopedia, Traumatologia, Rehabilitacja, 5(1), 80-85.

Earnshaw, S. R., & Dennett, S. L. (2003). Integer/linear mathematical programming models. Pharmacoeconomics, 21(12), 839-851.

Ela, G. K. (2004). Epidemiology of wilderness search and rescue in New Hampshire, 1999–2001. Wilderness & environmental medicine, 15(1), 11-17.

Epstein, D. M., Chalabi, Z., Claxton, K., & Sculpher, M. (2007). Efficiency, Equity, and

Budgetary Policies Informing Decisions Using Mathematical Programming. Medical Decision Making, 27(2), 128-137.

Flessa, S. (2003). Priorities and allocation of health care resources in developing countries: A case-study from the Mtwara region, Tanzania. European Journal of Operational Research, 150(1), 67-80.

Foissac, M., Millet, G. Y., Geyssant, A., Freychat, P., & Belli, A. (2009). Characterization of the mechanical properties of backpacks and their influence on the energetics of walking. Journal of biomechanics, 42(2), 125-130.

Fok, J., Adeeb, S., & Carey, J. (2010). FEM simulation of non-progressive growth from asymmetric loading and vicious cycle theory: scoliosis study proof of concept. Open Biomedical Engineering Journal, 4, 162-169.

Forjuoh, S. N., Little, D., Schuchmann, J. A., & Lane, B. L. (2003). Parental knowledge of school backpack weight and contents. Archives of disease in childhood, 88(1), 18-19.

Fowler, N. E., Rodacki, A. L., & Rodacki, C. D. (2006). Changes in stature and spine kinematics during a loaded walking task. Gait & posture, 23(2), 133-141.

Full, R. J., & Koditschek, D. E. (1999). Templates and anchors: neuromechanical hypotheses of legged locomotion on land. Journal of Experimental Biology, 202(23), 3325-3332.

Fusco, C., Zaina, F., Atanasio, S., Romano, M., Negrini, A., & Negrini, S. (2011). Physical exercises in the treatment of adolescent idiopathic scoliosis: An updated systematic review. Physiotherapy Theory and Practice, 27(1), 80-114.

Gagnon, D., Lariviere, C., & Loisel, P. (2001). Comparative ability of EMG, optimization, and hybrid modeling approaches to predict trunk muscle forces and lumbar spine loading during dynamic sagittal plane lifting. Clinical Biomechanics, 16, 359-372.

Gagnon, D., Arjmand, N., Plamondon, A., Shirazi-Adl, A., & Lariviere, C. (2011). An improved multi-joint EMG-assisted optimization approach to estimate joint and muscle forces in a musculoskeletal model of the lumbar spine. Journal of Biomechanics, 44, 1521-1529.

Ghandforoush, P. (1993). Optimal allocation of time in a hospital pharmacy using goal programming. European journal of operational research, 70(2), 191-198.

Ghufran, S., Khowaja, S., & Ahsan, M. J. (2015). Optimum multivariate stratified double sampling design: Chebyshev's Goal Programming approach. Journal of Applied Statistics, 42(5), 1032-1042.

Glassman, S. D., Bridwell, K., Dimar, J. R., Horton, W., Berven, S., & Schwab, F. (2005). The impact of positive sagittal balance in adult spinal deformity. Spine, 30(18), 2024-2029.

Goh, J. H., Thambyah, A., & Bose, K. (1998). Effects of varying backpack loads on peak forces in the lumbosacral spine during walking. Clinical Biomechanics, 13(1), S26-S31.

Golriz, S., & Walker, B. (2011). Can load carriage system weight, design and placement affect pain and discomfort? A systematic review. Journal of back and musculoskeletal rehabilitation, 24(1), 1-16.

Goodgold, S., Corcoran, M., Gamache, D., Gillis, J., Guerin, J., & Coyle, J. Q. (2002). Backpack use in children. Pediatric Physical Therapy, 14(3), 122-131.

Granata, K. P., & Marras, W. S. (1995a). An EMG-assisted model of trunk loading during free-dynamic lifting. Journal of Biomechanics, 28(11), 1309-1317.

Granata, K. P., & Marras, W. S. (1995b). The influence of trunk muscle coactivity on dynamic spinal load. Spine, 20(8), 913-919.

Grimmer, K., Dansie, B., Milanese, S., Pirunsan, U., & Trott, P. (2002). Adolescent standing postural response to backpack loads: a randomised controlled experimental study. BMC Musculoskeletal Disorders, 3(1), 10.

Grimmer, K., & Williams, J. (2003). Injury in junior Australian Rules footballers. Journal of Science and Medicine in sport, 6(3), 328-338.

Hajihosseinali, M., Arjmand, N., Shirazi-Adl, A., Farahmand, F., & Ghiasi, M. S. (2014). A novel stability and kinematics-driven trunk biomechanical model to estimate muscle and spinal forces. Medical Engineering and Physics, 36(10), 1296-1304.

Hale, C. J., Coleman, F. R., & Karpovich, P. V. (1953). Trunk inclination in carrying low and high packs of various weights (No. QMEPD-216). Springfield Coll., MA.

Hamzat, T. K., Abdulkareem, T. A., Akinyinka, O. O., & Fatoye, F. A. (2014). Backpack-related musculoskeletal symptoms among Nigerian secondary school students. Rheumatology international, 34(9), 1267-1273.

Hamonko, M. T., McIntosh, S. E., Schimelpfenig, T., & Leemon, D. (2011). Injuries related to hiking with a pack during National Outdoor Leadership School courses: A risk factor analysis. Wilderness & environmental medicine, 22(1), 2-6.

Hande, D. N., Shinde, N., Khatri, S. M., & Dangat, P. (2012). The Effect of Backpack on Cervical and Shoulder Posture in Male Students of Loni. International Journal of

Health Sciences & Research, 2(3), 72-79.

Harman, E., Hoon, K., Frykman, P., & Pandorf, C. (2000). The effects of backpack weight on the biomechanics of load carriage (No. USARIEM-T00-17). Army Research Inst. of Environmental Medicine, Natick MA Military Performance Div.

Haselgrove, C., Straker, L., Smith, A., O'Sullivan, P., Perry, M., & Sloan, N. (2008). Perceived school bag load, duration of carriage, and method of transport to school are associated with spinal pain in adolescents: an observational study. Australian Journal of Physiotherapy, 54(3), 193-200.

Heller, M. F., Challis, J. H., & Sharkey, N. A. (2009). Changes in postural sway as a consequence of wearing a military backpack. Gait & posture, 30(1), 115-117.

Hendershot, B. D., & Wolf, E. J. (2014). Three-dimensional joint reaction forces and moments at the low back during over-ground walking in persons with unilateral lower-extremity amputation. Clinical Biomechanics, 29(3), 235-242.

Heuscher, Z., Gilkey, D. P., Peel, J. L., & Kennedy, C. A. (2010). The association of self-reported backpack use and backpack weight with low back pain among college students. Journal of manipulative and physiological therapeutics, 33(6), 432-437.

Holewijin, M. (1990). “Physiological stain due to load carrying.” European Jouranl of Applied Phisiology and Occupational Phsiology, 61:237-245.

Holt, K. G., Wagenaar, R. C., LaFiandra, M. E., Kubo, M., & Obusek, J. P. (2003). Increased musculoskeletal stiffness during load carriage at increasing walking speeds maintains constant vertical excursion of the body center of mass. Journal of biomechanics, 36(4), 465-471.

Hong, Y., & Cheung, C. K. (2003). Gait and posture responses to backpack load during level walking in children. Gait & posture, 17(1), 28-33.

Hong, Y., Li, J. X., & Fong, D. T. P. (2008). Effect of prolonged walking with backpack loads on trunk muscle activity and fatigue in children. Journal of Electromyography and Kinesiology, 18(6), 990-996.

Hong, Y., Li, J. X., Wong, A. S. K., & Robinson, P. D. (2000). Effects of load carriage on heart rate, blood pressure and energy expenditure in children. Ergonomics, 43(6), 717-727.

Hough, P. A., Nel, M., Smit, J. E., Malan, E., van der Watt, M., Deacon, A. F., ... & Bester, A. M. (2006). The Influence of carrying a school bag on the developing spine. Children's Health Care, 35(4), 339-348.

Hresko, M. T. (2013). Idiopathic scoliosis in adolescents. New England Journal of Medicine, 368(9), 834-841.

Hu, B., Shan, X., Zhou, J., & Ning, X. (2014). The effects stance width and foot pressure on lumbar muscle flexion-relaxation phenomenon. Clinical Biomechanics, 29:311-316.

Hung, W. L., Lee, E. S., & Chuang, S. C. (2011). Balanced bootstrap resampling method for neural model selection. Computers and Mathematics with Applications, 62(12), 4576-4581.

Ibrahim, A. H. (2012). Incidence of Back Pain in Egyptian School Girls: Effect of School Bag Weight and Carrying Waymu. World applied sciences journal, 17(11), 1526-1534.

Ikeda, E. R., Cooper, L., Gulick, P., & Nguyen, P. (2008). The metabolic cost of carrying a single-versus double-strap golf bag. The Journal of Strength & Conditioning Research, 22(3), 974-977.

Ignizio, J. P. (1978a). A review of goal programming: A tool for multiobjective analysis. Journal of the Operational Research Society, 1109-1119.

Ignizio, J. P. (1978b). The development of cost estimating relationships via goal programming. The Engineering Economist, 24(1), 37-48.

Imrie, M., Yaszay, B., Bastrom, T. P., Wenger, D. R., & Newton, P. O. (2011). Adolescent idiopathic scoliosis: Should 100% correction be the goal? Journal of Pediatric Orthopaedics, 31, S13.

Ishida, H., Watanabe, S., Yanagawa, H., Kawasaki, M., Kobayashi, Y., & Amano, Y. (2008). Immediate effects of a rucksack type orthosis on the elderly with decreased lumbar lordosis during standing and walking. Electromyography and clinical neurophysiology, 48(1), 53-61.

Iyer, S. R. (2001). An ergonomic study of chronic musculoskeletal pain in schoolchildren. Indian journal of pediatrics, 68(10), 937-941.

Jacobsen,G. (2009). Police buckle under pressure. The Sydney Morning Herald. http://www.smh.com.au/national/police-buckle-under-pressure-20090308-8sh2?deviceType=text

Janakiraman, B., Ravichandran, H., Demeke, S., & Fasika, S. (2017). Reported influences of backpack loads on postural deviation among school children: A systematic review. Journal of Education and Health Promotion, 6(1), 41.

Jayaratne, K. (2012). Inculcating the ergonomic culture in developing countries: national healthy schoolbag initiative in Sri Lanka. Human factors, 54(6), 908-924.

Jones, D., & Tamiz, M. (2010). Practical goal programming (Vol. 141). New York: Springer.

Jonsson, B. (1982). Measurement and evaluation of local muscular strain in the shoulder during constrained work. Journal of human ergology, 11(1), 73-88.

Kamran, M. A., Karimi, B., Bakhtiari, H., & Masoumzadeh, S. (2017). A resource allocation model in a healthcare emergency center using goal programming. Journal of Engineering Research, 4(4).

Kazemzadeh, R. B., Bashiri, M., Atkinson, A. C., & Noorossana, R. (2008). A general framework for multiresponse optimization problems based on goal programming. European Journal of Operational Research, 189(2), 421-429.

Kennedy, C., Kassab, O., Gilkey, D., Linnel, S., & Morris, D. (2008). Psychosocial factors and low back pain among college students. Journal of American College Health, 57(2), 191-196.

Keown, A. J., & Taylor, B. W. (1978). Integer goal programming model for the implementation of multiple corporate objectives. Journal of Business Research, 6(3), 221-235.

Khoo, B. C. C., Goh, J. C. H., & Bose, K. (1995). A biomechanical model to determine lumbosacral loads during single stance phase in normal gait. Medical engineering & physics, 17(1), 27-35.

Khorramshahgol, R., Tamiz, M., & Al-Hussain, R. (2014). Application of goal programming to swap analysis in oil industry. Journal of Information and Optimization Sciences, 35(1), 73-91.

Kim, D. R., Ali, M., Sur, D., Khatib, A., & Wierzba, T. F. (2012). Determining optimal neighborhood size for ecological studies using leave-one-out cross validation. International Journal of Health Geographics, 11(1), 10.

Kim, M. H., Yi, C. H., Kwon, O. Y., Cho, S. H., & Yoo, W. G. (2008). Changes in neck muscle electromyography and forward head posture of children when carrying schoolbags. Ergonomics, 51(6), 890-901.

Kinoshita, H. (1985). Effects of different loads and carrying systems on selected biomechanical parameters describing walking gait. Ergonomics, 28(9), 1347-1362.

Kirk, J., & Schneider, D. A. (1992). Physiological and perceptual responses to load–carrying in female subjects using internal and external frame backpacks. Ergonomics, 35(4), 445-455.

Kistner, F., Fiebert, I., & Roach, K. (2012). Effect of backpack load carriage on cervical posture in primary schoolchildren. Work, 41(1), 99-108.

Knapik, J., Harman, E., & Reynolds, K. (1996). Load carriage using packs: a review of physiological, biomechanical and medical aspects. Applied ergonomics, 27(3), 207-216.

Knapik, J. J., Reynolds, K. L., & Harman, E. (2004). Soldier load carriage: historical, physiological, biomechanical, and medical aspects. Military medicine, 169(1), 45.

Korovessis, P., Koureas, G., & Papazisis, Z. (2004). Correlation between backpack weight and way of carrying, sagittal and frontal spinal curvatures, athletic activity, and dorsal and low back pain in schoolchildren and adolescents.

Clinical Spine Surgery, 17(1), 33-40.

Korovessis, P., Koureas, G., Zacharatos, S., & Papazisis, Z. (2005). Backpacks, back pain, sagittal spinal curves and trunk alignment in adolescents: a logistic and multinomial logistic analysis. Spine, 30(2), 247-255.

Kwak, N. K., & Lee, C. W. (2002). Business process reengineering for health-care system using multicriteria mathematical programming. European Journal of Operational Research, 140(2), 447-458.

LaFiandra, M., Wagenaar, R. C., Holt, K. G., & Obusek, J. P. (2003). How do load carriage and walking speed influence trunk coordination and stride parameters?. Journal of biomechanics, 36(1), 87-95.

Lafortune, P., Aubin, C. É., Boulanger, H., Villemure, I., Bagnall, K. M., & Moreau, A. (2007). Biomechanical simulations of the scoliotic deformation process in the pinealectomized chicken: a preliminary study. Scoliosis, 2(1), 16.

Lafortune, D., Norman, R. W., & McGill, S. M. (1988, August). Ensemble average of linear enveloped EMG's during lifting. In Proceedings of the Biannual Conference of the Canadian Society for Biomechanics, Ottawa, Canada.

Lai, E. S. Y., Chow, D. K. M., Tsang, V. N. F., Tsui, S. S. I., Lam, C. Y., Su, I. Y. W., & Chow, D. H. (2011). Effect of load carriage on chronic low back pain in adults with cerebral palsy. Prosthetics and orthotics international, 35(4), 439-444.

Lavigne, V. (2014). Weight limit recommendation in backpack use for school-aged children. J Clin Chirop Pediatr, 14(2), 1156-1159.

Leboeuf-Yde, C., & Kyvik, K. O. (1998). At what age does low back pain become a common problem?. A study of 29,424 individuals aged 12-41 Years. Spine, 23(2), 228-234.

Leboeuf-Yde, C., Fejer, R., Nielsen, J., Kyvik, K. O., & Hartvigsen, J. (2012). Pain in the three spinal regions: the same disorder? Data from a population-based sample of 34,902 Danish adults. Chiropractic & manual therapies, 20(1), 11.

Lee, C. W., & Kwak, N. K. (1999). Information resource planning for a health-care system using an AHP-based goal programming method. Journal of the operational research society, 50(12), 1191-1198.

Lee, C. W., & Kwak, N. K. (2011). Strategic enterprise resource planning in a health-care system using a multicriteria decision-making model. Journal of medical systems, 35(2), 265-275.

Lee, S. M., & Lerro, A. J. (1974). Capital budgeting for multiple objectives. Financial Management, 58-66.

Legg, S. J. (1985). Comparison of different methods of load carriage. Ergonomics, 28(1), 197-212.

Legg, S. J., & Cruz, C. O. (2004). Effect of single and double strap backpacks on lung function. Ergonomics, 47(3), 318-323.

Legg, S. J., & Mahanty, A. (1985). Comparison of five modes of carrying a load close to the trunk. Ergonomics, 28(12), 1653-1660.

Legg, S. J., Perko, L., & Campbell, P. (1997). Subjective perceptual methods for comparing backpacks. Ergonomics, 40(8), 809-817.

Legg, S. J., Ramsey, T., & Knowles, D. J. (1992). The metabolic cost of backpack and shoulder load carriage. Ergonomics, 35(9), 1063-1068.

Lenhoff, M. W., Santner, T. J., Otis, J. C., Peterson, M. G., Williams, B. J., & Backus, S. I.

(1999). Bootstrap prediction and confidence bands: a superior statistical method for analysis of gait data. Gait Posture, 9(1), 10-17.

Li, X., & Aruin, A. S. (2009). The effect of short-term changes in body mass distribution on feed-forward postural control. Journal of Electromyography and Kinesiology, 19(5), 931-941.

Li, S. S. W., & Chow, D. H. K. (2018). Effects of backpack load on critical changes of trunk muscle activation and lumbar spine loading during walking. Ergonomics, 61(4), 553-565.

Li, S. S. W., & Chow, D. H. K. (2016). Multi-objective analysis for assessing simultaneous changes in regional spinal curvatures under backpack carriage in young adults. Ergonomics, 59(11), 1494-1504.

Li, J. X., Hong, Y., & Robinson, P. D. (2003). The effect of load carriage on movement kinematics and respiratory parameters in children during walking. European Journal of Applied Physiology, 90(1-2), 35-43.

Liberatore, M. J., & Nydick, R. L. (2008). The analytic hierarchy process in medical and health care decision making: A literature review. European Journal of Operational Research, 189(1), 194-207.

Liew, B., Morris, S., & Netto, K. (2015). The effect of backpack carriage on the biomechanics and neuromuscular control of locomotion: A systematic review and preliminary meta-analysis. Journal of Science and Medicine in Sport, (19), e42-e43.

Lindstrom-Hazel, D. (2009). The backpack problem is evident but the solution is less obvious. Work, 32(3), 329-338.

Little, D. G., Song, K. M., Katz, D., & Herring, J. A. (2000). Relationship of peak height velocity to other maturity indicators in idiopathic scoliosis in girls. The Journal of Bone & Joint Surgery, 82(5), 685-685.

Lloyd, R., & Cooke, C. B. (2000). The oxygen consumption with unloaded walking and load carriage using two different backpack designs. European Journal of Applied Physiology, 81(6), 486-492.

Lloyd, R., Parr, B., Davies, S., & Cooke, C. (2010). Subjective perceptions of load carriage on the head and back in Xhosa women. Applied ergonomics, 41(4), 522-529.

Lobb, B. (2004). Load carriage for fun: a survey of New Zealand trampers, their activities and injuries. Applied ergonomics, 35(6), 541-547.

Lucas, G. N. (2011). Backpacks in children. Sri Lanka Journal of Child Health, 40:1-3.

Luk, K. D., Lee, C. F., Cheung, K. M., Cheng, J. C., Ng, B. K., Lam, T. P., . . . Fong, D. Y. (2010). Clinical effectiveness of school screening for adolescent idiopathic scoliosis: A large population-based retrospective cohort study. Spine, 35(17), 1607-1614.

Luk, K. D. K., Yip, D. K. H., & Tse, L. Y. (1997). Schoolchildren screening for scoliosis in Hong Kong. In 17th Annual Congress of the Hong Kong Orthopaedic Association, November 15-16, 1997.

Macias, B. R., Murthy, G., Chambers, H., & Hargens, A. R. (2008). Asymmetric loads and pain associated with backpack carrying by children. Journal of Pediatric Orthopaedics, 28(5), 512-517.

Macias, B. R., Murthy, G., Chambers, H., & Hargens, A. R. (2005). High contact pressure beneath backpack straps of children contributes to pain. Archives of pediatrics & adolescent medicine, 159(12), 1186-1188.

Mackie, H. W., & Legg, S. J. (2007). Measurement of the temporal patterns of school bag carriage using activity monitoring and structured interview. Ergonomics, 50(10), 1668-1679.

Mackie, H. W., & Legg, S. J. (2008). Postural and subjective responses to realistic schoolbag carriage. Ergonomics, 51(2), 217-231.

Mac-Thiong, J. M., Labelle, H., & Roussouly, P. (2011). Pediatric sagittal alignment. European Spine Journal, 20(5), 586.

Mac-Thiong, J. M., Transfeldt, E. E., Mehbod, A. A., Perra, J. H., Denis, F., Garvey, T. A., ... & Winter, R. B. (2009). Can c7 plumbline and gravity line predict health related quality of life in adult scoliosis?. Spine, 34(15), E519-E527.

Madras, D. E., Cornwall, M. W., & Coast, J. R. (1998). Energy cost, perceived exertion and postural adjustments when treadmill walking with two types of backpack. Journal of Human Movement Studies, 35(5), 233-249.

Mahoney, C. R., Hirsch, E., Hasselquist, L., Lesher, L. L., & Lieberman, H. R. (2007). The effects of movement and physical exertion on soldier vigilance. Aviation, space, and environmental medicine, 78(5), B51-B57.

Majumdar, D., Pal, M. S., & Majumdar, D. (2010). Effects of military load carriage on kinematics of gait. Ergonomics, 53(6), 782-791.

Majumdar, D., Pal, M. S., Pramanik, A., & Majumdar, D. (2013). Kinetic changes in gait during low magnitude military load carriage. Ergonomics, 56(12), 1917-1927.

Malhotra, M. S., & Gupta, J. S. (1965). Carrying of school bags by children. Ergonomics, 8(1), 55-60.

Maloiy, G. M., Heglund, N. C., Prager, L. M., Cavagna, G. A., & Taylor, C. R. (1986). Energetic cost of carrying loads: have African women discovered an economic way?. Nature, 319(6055), 668-669.

Mamyama, T., Kitagawal, T., Takeshita, K., & Nakainura, K. (2001). Side shift exercise for idiopathic scoliosis after skeletal maturity. Studies in Health Technology and Informatics, 91, 361-364.

Marras, W. S., & Granata, K. P. (1997). The development of an EMG-assisted model to assess spine loading during whole-body free-dynamic lifting. Journal of Electromyography Kinesiology, 7(4), 259-268.

Martel, J. M., & Aouni, B. (1990). Incorporating the decision-maker's preferences in the goal-programming model. Journal of the Operational Research Society, 1121-1132.

Maruyama, T., Kitagawa, T., Takeshita, K., Mochizuki, K., & Nakamura, K. (2003). Conservative treatment for adolescent idiopathic scoliosis: Can it reduce the incidence of surgical treatment? Pediatric Rehabilitation, 6(3-4), 215-219.

May, B. (2009). Effects of backpack load on balance and decisional processes. Military medicine, 174(12), 1308.

Mayo, N. E., Goldberg, M. S., Poitras, B., Scott, S., & Hanley, J. (1994). The Ste-justine adolescent idiopathic scoliosis cohort study: Part III: Back pain. Spine, 19(14), 1573-1581.

McGill, S. M. 1992. A myoelectrically based dynamic three-dimensional model to predict loads on lumbar spine tissue during lateral bending. Journal of Biomechanics, 26(4), 395-414.

McGill, S. M., Marshall, L., & Andersen, J. (2013). Low back loads while walking and carrying: comparing the load carried in one hand or in both hands. Ergonomics, 56(2), 293-302.

McGill, S. M., & Norman, R. W. (1986). Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. Spine, 11(7), 666-678.

McIlroy, W. E., & Maki, B. E. (1997). Preferred placement of the feet during quiet stance: development of a standardized foot placement for balance testing. Clinical Biomechanics, 12(1), 66-70.

Meakin, J. R., Gregory, J. S., Smith, F. W., Gilbert, F. J., & Aspden, R. M. (2008). Characterizing the shape of the lumbar spine using an active shape model: reliability and precision of the method. Spine, 33(7), 807-813.

Mikshowsky, A. A., Gianola, D., & Weigel, K. A. (2017). Assessing genomic prediction accuracy for Holstein sires using bootstrap aggregation sampling and leave-one-out cross validation. Journal of Dairy Science, 100(1), 453-464.

Minetti, A. E., Formenti, F., & Ardigò, L. P. (2006). Himalayan porter's specialization: metabolic power, economy, efficiency and skill. Proceedings of the Royal Society of London B: Biological Sciences, 273(1602), 2791-2797.

Minghelli, B., Oliveira, R., & Nunes, C. (2014). Non-specific low back pain in adolescents from the south of Portugal: prevalence and associated factors.

Journal of Orthopaedic Science, 19(6), 883-892.

Mohammadi, Y., Arjmand, N., & Shirazi-Adl, A. (2015). Comparison of trunk muscle forces, spinal loads and stability estimated by one stability- and three EMG-assisted optimization approaches. Medical Engineering and Physics, 37(8), 792-800.

Mooney, V., & Brigham, A. (2003). The role of measured resistance exercises in adolescent scoliosis. Orthopedics, 26(2), 167-171.

Mooney, V., Gulick, J., & Pozos, R. (2000). A preliminary report on the effect of measured strength training in adolescent idiopathic scoliosis. Clinical Spine Surgery, 13(2), 102-107.

Moore, M. J., White, G. L., & Moore, D. L. (2007). Association of relative backpack weight with reported pain, pain sites, medical utilization, and lost school time in children and adolescents. Journal of School Health, 77(5), 232-239.

Morton, A., Keisler, J. M., & Salo, A. (2016). Multicriteria portfolio decision analysis for project selection. In Multiple Criteria Decision Analysis (pp. 1269-1298). Springer New York.

Motmans, R. R. E. E., Tomlow, S., & Vissers, D. (2006). Trunk muscle activity in different modes of carrying schoolbags. Ergonomics, 49(2), 127-138.

Muirhead, A., & Conner, A. N. (1985). The assessment of lung function in children with scoliosis. Bone & Joint Journal, 67(5), 699-702.

Mwaka, E. S., Munabi, I. G., Buwembo, W., Kukkiriza, J., & Ochieng, J. (2014). Musculoskeletal pain and school bag use: a cross-sectional study among Ugandan pupils. BMC research notes, 7(1), 222.

Narasimhan, R. (1980). Goal programming in a fuzzy environment. Decision sciences, 11(2), 325-336.

Nault, M., Allard, P., Hinse, S., Le Blanc, R., Caron, O., Labelle, H., & Sadeghi, H. (2002). Relations between standing stability and body posture parameters in adolescent idiopathic scoliosis. Spine, 27(17), 1911-1917.

Navuluri, N., & Navuluri, R. B. (2006). Study on the relationship between backpack use and back and neck pain among adolescents. Nursing & health sciences, 8(4), 208-215.

Negrini, S., Aulisa, A. G., Aulisa, L., Circo, A. B., de Mauroy, J. C., Durmala, J., Maruyama, T. (2012). 2011 SOSORT guidelines: Orthopaedic and rehabilitation treatment of idiopathic scoliosis during growth. Scoliosis, 7(1), 3.

Negrini, S., & Carabalona, R. (2002). Backpacks on! Schoolchildren's perceptions of load, associations with back pain and factors determining the load. Spine, 27(2), 187-195.

Negrini, S., Carabalona, R., & Sibilla, P. (1999). Backpack as a daily load for schoolchildren. The Lancet, 354(9194), 1974.

Negrini, S., & Negrini, A. (2007). Postural effects of symmetrical and asymmetrical loads on the spines of schoolchildren. Scoliosis, 2(1), 8.

Negrini, S., Politano, E., Carabalona, R., Tartarotti, L., & Marchetti, M. L. (2004). The backpack load in schoolchildren: clinical and social importance, and efficacy of a community-based educational intervention. Europa medicophysica.

Nie, W., Ye, M., & Wang, Z. (2008). Infinite models in scoliosis: A review of the literature and analysis of personal experience. Biomedizinische

Technik/Biomedical Engineering, 53(4), 174-180.

Nussbaum, M. A., & Chaffin, D. B. (1998). Lumbar muscle force estimation using a subject-invariant 5-parameter EMG-based model. Journal of Biomechanics, 31(7), 667-672.

Nussbaum, M. A., Chaffin, D. B., & Baker, G. (1999). Biomechanical analysis of materials handling manipulators in short distance transfers of moderate mass objects: joint strength, spine forces and muscular antagonism. Ergonomics, 42(12), 1597-1618.

Nutter, P. (1988). Aerobic exercise in the treatment and prevention of low back pain. Occupational medicine (Philadelphia, Pa.), 3(1), 137-145.

Oddoye, J. P. (2006). Efficient resource planning in a medical assessment unit using goal programming and simulation (Doctoral dissertation, University of Portsmouth).

Oddoye, J. P., Jones, D. F., Tamiz, M., & Schmidt, P. (2009). Combining simulation and goal programming for healthcare planning in a medical assessment unit. European Journal of Operational Research, 193(1), 250-261.

Olmedo-Buenrostro, B. A., Delgado-Enciso, I., Sánchez-Ramírez, C. A., Cruz, S. A. M., Vásquez, C., Mora-Brambila, A. B., ... & Martínez-Fierro, M. L. (2016). Current Backpack Weight Status for Primary Schoolchildren in Colima, Mexico. The Journal of School Nursing, 32(3), 172-176.

Onofrio, A. C., Da Silva, M. C., Domingues, M. R., & Rombaldi, A. J. (2012). Acute low back pain in high school adolescents in Southern Brazil: prevalence and associated factors. European Spine Journal, 21(7), 1234-1240.

Orloff, H. A., & Rapp, C. M. (2004). The effects of load carriage on spinal curvature and posture. Spine, 29(12), 1325-1329.

Orr, R. M., Johnston, V., Coyle, J., & Pope, R. (2015). Reported load carriage injuries of the Australian army soldier. Journal of occupational rehabilitation, 25(2), 316-322.

Ozgül, B., Akalan, N. E., Kuchimov, S., Uygur, F., Temelli, Y., & Polat, M. G. (2012). Effects of unilateral backpack carriage on biomechanics of gait in adolescents: a kinematic analysis. Acta Orthop Traumatol Turc, 46(4), 269-274.

Pakzad, M., Fung, J., & Preuss, R. (2016). Pain catastrophizing and trunk muscle activation during walking in patients with chronic low back pain. Gait & posture, 49, 73-77.

Pal, G. P. (1991). Mechanism of Production of Scoliosis: A Hypothesis. Spine, 16(3), 288-292.

Palmer, C. J., Bigelow, C., & Van Emmerik, R. E. (2013). Defining soldier equipment trade space: load effects on combat marksmanship and perception–action coupling. Ergonomics, 56(11), 1708-1721.

Park, K., Hur, P., Rosengren, K. S., Horn, G. P., & Hsiao-Wecksler, E. T. (2010). Effect of load carriage on gait due to firefighting air bottle configuration. Ergonomics, 53(7), 882-891.

Pascoe, D. D., Pascoe, D. E., Wang, Y. T., Shim, D. M., & Kim, C. K. (1997). Influence of carrying book bags on gait cycle and posture of youths. Ergonomics, 40(6), 631-640.

Pau, M., & Pau, M. (2010). Postural sway modifications induced by backpack carriage in primary school children: a case study in Italy. Ergonomics, 53(7), 872-881.

Paušić, J., Kujundžić, H., & Penjak, A. (2013). Possible influences of a heavy backpack on a painful back pain syndrome in elementary school children. Hrvatski časopis za odgoj i obrazovanje, 15(2), 275-287.

Pierrynowski, M. R., Winter, D. A., & Norman, R. W. (1981). Metabolic measures to ascertain the optimal load to be carried by man. Ergonomics, 24(5), 393-399.

Piscione, J., & Gamet, D. (2006). Effect of mechanical compression due to load carrying on shoulder muscle fatigue during sustained isometric arm abduction: an electromyographic study. European journal of applied physiology, 97(5), 573-581.

Plamondon, A., Gagnon, M., & Desjardins, P. (1996). Validation of two 3-D segment models to calculate the net reaction forces and moments at the L5/S1 joint in lifting. Clinical Biomechanics, 11(2), 101-110.

Puckree, T., Silal, S. P., & Lin, J. (2004). School bag carriage and pain in school children. Disability and Rehabilitation, 26(1), 54-59.

Ramprasad, M., Alias, J., & Raghuveer, A. K. (2010). Effect of backpack weight on postural angles in preadolescent children. Indian pediatrics, 47(7), 575-580.

Rateau, M. R. (2004). Use of backpacks in children and adolescents: a potential contributor of back pain. Orthopaedic nursing, 23(2), 101-105.

Reed, M. P., Manary, M. A., & Schneider, L. W. (1999). Methods for measuring and representing automobile occupant posture (No. 1999-01-0959). SAE Technical Paper.

Reeves, N. P., Narendra, K. S., & Cholewicki, J. (2007). Spine stability: the six blind men and the elephant. Clinical Biomechanics, 22(3), 266-274.

Reeves, N. P., Narendra, K. S., & Cholewicki, J. (2011). Spine stability: lessons from balancing a stick. Clinical Biomechanics, 26(4), 325-330.

Roaf, R. (1966). The basic anatomy of scoliosis. J Bone Joint Surg Br, 48(4), 786-792.

Rodríguez-Soto, A. E., Jaworski, R., Jensen, A., Niederberger, B., Hargens, A. R., Frank, L. R., ... & Ward, S. R. (2013). Effect of load carriage on lumbar spine kinematics. Spine, 38(13), E783-E791.

Romero, C. (1984). A note—Effects of five sided penalty functions in goal programming. Omega, 12(4), 333.

Roussouly, P., & Pinheiro-Franco, J. L. (2011). Sagittal parameters of the spine: biomechanical approach. European Spine Journal, 20(5), 578.

Rose, J. D., Mendel, E., & Marras, W. S. (2013). Carrying and spine loading. Ergonomics, 56(11), 1722-1732.

Rugelj, D., & Sevšek, F. (2011). The effect of load mass and its placement on postural sway. Applied ergonomics, 42(6), 860-866.

Sahlstrand, T., Ortengren, R., & Nachemson, A. (1978). Postural equilibrium in adolescent idiopathic scoliosis. Acta Orthopaedica Scandinavica, 49(4), 354-365.

Seay, J. F. (2015). Biomechanics of Load Carriage—Historical Perspectives and Recent Insights. The Journal of Strength & Conditioning Research, 29, S129-S133.

Sevastir, J. A., Aaro, S., & Normelli, H. (1984). Scoliosis: Experimental and Clinical Studies. Clinical orthopaedics and related research, 191, 27-34.

Shamsoddini, A. R., Hollisaz, M. T., & Hafezi, R. (2010). Backpack weight and musculoskeletal symptoms in secondary school students, Tehran, Iran. Iranian journal of public health, 39(4), 120.

Sharkey, & Gaskill. (2009).National Wildfire Coordinating Group.Fitness and Work Capacity, 2009 Edition, PMS 304–2. Available at: http://www.nwcg.gov/pms/pubs/pms304-2.pdf.

Sheir-Neiss, G. I., Kruse, R. W., Rahman, T., Jacobson, L. P., & Pelli, J. A. (2003). The association of backpack use and back pain in adolescents. Spine, 28(9), 922-930.

Shi, L., Wang, D., Driscoll, M., Villemure, I., Chu, W. C., Cheng, J. C., & Aubin, C. (2011). Biomechanical analysis and modeling of different vertebral growth patterns in adolescent idiopathic scoliosis and healthy subjects. Scoliosis, 6(1), 11.

Shojaei, I., Hendershot, B. D., Wolf, E. J., & Bazrgari, B. (2016). Persons with unilateral transfemoral amputation experience larger spinal loads during level-ground walking compared to able-bodied individuals. Clinical Biomechanics, 32, 157-163.

Siambanes, D., Martinez, J. W., Butler, E. W., & Haider, T. (2004). Influence of school backpacks on adolescent back pain. Journal of Pediatric Orthopaedics, 24(2), 211-217.

Simpson, K. M., Munro, B. J., & Steele, J. R. (2011a). Backpack load affects lower limb muscle activity patterns of female hikers during prolonged load carriage.

Journal of Electromyography and Kinesiology, 21(5), 782-788.

Simpson, K. M., Munro, B. J., & Steele, J. R. (2011b). Effect of load mass on posture, heart rate and subjective responses of recreational female hikers to prolonged load carriage. Applied ergonomics, 42(3), 403-410.

Simpson, K. M., Munro, B. J., & Steele, J. R. (2012a). Does load position affect gait and subjective responses of females during load carriage?. Applied ergonomics, 43(3), 479-485.

Simpson, K. M., Munro, B. J., & Steele, J. R. (2012b). Effects of prolonged load carriage on ground reaction forces, lower limb kinematics and spatio-temporal parameters in female recreational hikers. Ergonomics, 55(3), 316-326.

Singh, T., & Koh, M. (2009). Effects of backpack load position on spatiotemporal parameters and trunk forward lean. Gait & posture, 29(1), 49-53.

Smith, B., Ashton, K. M., Bohl, D., Clark, R. C., Metheny, J. B., & Klassen, S. (2006). Influence of carrying a backpack on pelvic tilt, rotation, and obliquity in female college students. Gait & posture, 23(3), 263-267.

Steiger, J. H. (2000). Point estimation, hypothesis testing, and interval estimation using the RMSEA: Some comments and a reply to Hayduk and Glaser. Structural Equation Modeling, 7(2), 149-162.

Stokes, I. A. (2007). Analysis and simulation of progressive adolescent scoliosis by biomechanical growth modulation. European Spine Journal, 16(10), 1621-1628.

Stokes, I. A. F. (2002). Mechanical effects on skeletal growth. Journal of Musculoskeletal and Neuronal Interactions, 2(3), 277-280.

Stokes, I. A. (1994). Three-Dimensional Terminology of Spinal Deformity: A Report Presented to the Scoliosis Research Society by The Scoliosis Research Society Working Group on 3-D Terminology of Spinal Deformity. Spine, 19(2), 236-248.

Stokes, I. A., Burwell, R. G., & Dangerfield, P. H. (2006). Biomechanical spinal growth modulation and progressive adolescent scoliosis–a test of the'vicious cycle'pathogenetic hypothesis: Summary of an electronic focus group debate of the IBSE. Scoliosis, 1(1), 16.

Stokes, I. A., & Gardner-Morse, M. (2001). Lumbar spinal muscle activation synergies predicted by multi-criteria cost function. Journal of Biomechanics, 34(6), 733-740.

Stokes, I. A., Mente, P. L., Iatridis, J. C., Farnum, C. E., & Aronsson, D. D. (2002). Enlargement of growth plate chondrocytes modulated by sustained mechanical loading. J Bone Joint Surg Am, 84(10), 1842-1848.

Strube, E. M., Sumner, A., Kollock Jr, R. O., Games, K. E., Lackamp, M. A., Mizutani, M., & Sefton, J. M. (2017). The Effect of Military Load Carriage on Postural Sway, Forward Trunk Lean, and Pelvic Girdle Motion. International Journal of Exercise Science, 10(1), 3.

Stuempfle, K. J., Drury, D. G., & Wilson, A. L. (2004). Effect of load position on physiological and perceptual responses during load carriage with an internal frame backpack. Ergonomics, 47(7), 784-789.

Talbott, N. R., Bhattacharya, A., Davis, K. G., Shukla, R., & Levin, L. (2009). School backpacks: it's more than just a weight problem. Work, 34(4), 481-494.

Tamiz, M., Jones, D. F., & El-Darzi, E. (1995). A review of goal programming and its applications. Annals of Operations Research, 58(1), 39-53.

Tamiz, M., Jones, D., & Romero, C. (1998). Goal programming for decision making: An overview of the current state-of-the-art. European Journal of operational research, 111(3), 569-581.

Taylor III, B. W., & Keown, A. J. (1978). Planning urban recreational facilities with integer goal programming. Journal of the Operational Research Society, 751-758.

Thokala, P., & Duenas, A. (2012). Multiple criteria decision analysis for health technology assessment. Value in Health, 15(8), 1172-1181.

Tingley, K. M., & Liebman, J. S. (1984). A goal programming example in public health resource allocation. Management Science, 30(3), 279-289.

Trevelyan, F. C., & Legg, S. J. (2011). Risk factors associated with back pain in New Zealand school children. Ergonomics, 54(3), 257-262.

Troussier, B., Davoine, P., De Gaudemaris, R., Fauconnier, J., & Phelip, X. (1994). Back pain in school children. A study among 1178 pupils. Scandinavian journal of rehabilitation medicine, 26(3), 143-146.

Tzu-wei, P. H., & Kuo, A. D. (2014). Mechanics and energetics of load carriage during human walking. Journal of Experimental Biology, 217(4), 605-613.

Vacheron, J. J., Poumarat, G., Chandezon, R., & Vanneuville, G. (1999). Changes of contour of the spine caused by load carrying. Surgical and Radiologic Anatomy, 21(2), 109-113.

Van Dieen, J. H., & Kingma, I. (2005). Effects of antagonistic co-contraction on differences between electromyography based and optimization based estimates of spinal forces. Ergonomics, 48(4), 411-426.

Van Gent, C., Dols, J. J., Carolien, M., Sing, R. A. H., & de Vet, H. C. (2003). The weight of schoolbags and the occurrence of neck, shoulder, and back pain in young adolescents. Spine, 28(9), 916-921.

Vaughan, C., Davis, B., & O'Connor, J. C. (1999). Dynamics of human gait (2nd edition). Kiboho Publishers, Cape Town, South Africa, p. 17-23.

Vidal, J., Borràs, P. A., Ponseti, F. J., Cantallops, J., Ortega, F. B., & Palou, P. (2013). Effects of a postural education program on school backpack habits related to low back pain in children. European Spine Journal, 22(4), 782-787.

Vieira, A. C., & Ribeiro, F. (2015). Impact of backpack type on respiratory muscle strength and lung function in children. Ergonomics, 58(6), 1005-1011.

Villemure, I., Aubin, C. E., Dansereau, J., & Labelle, H. (2004). Biomechanical simulations of the spine deformation process in adolescent idiopathic scoliosis from different pathogenesis hypotheses. European Spine Journal, 13(1), 83-90.

Villemure, I., Aubina, C. E., Dansereau, J., & Labelle, H. (2002). Simulation of progressive deformities in adolescent idiopathic scoliosis using a biomechanical model integrating vertebral growth modulation. Journal of biomechanical engineering, 124(6), 784-790.

Von Heimburg, E. D., Rasmussen, A. K. R., & Medbø, J. I. (2006). Physiological responses of firefighters and performance predictors during a simulated rescue of hospital patients. Ergonomics, 49(2), 111-126.

Wagner, H., Liebetrau, A., Schinowski, D., Wulf, T., de Lussanet, M. H. (2012). Spinal lordosis optimizes the requirements for a stable erect posture.

Theoretical Biology and Medical Modelling, 9(1), 13.

Wall, E. J., Foad, S. L., & Spears, J. (2003). Backpacks and back pain: where's the epidemic?. Journal of Pediatric Orthopaedics, 23(4), 437-439.

Wang, W., Baran, G. R., Betz, R. R., Samdani, A. F., Pahys, J. M., & Cahill, P. J. (2014). The use of finite element models to assist understanding and treatment for scoliosis: A review paper. Spine Deformity, 2(1), 10-27.

Weinstein, S. L., Dolan, L. A., Cheng, J. C., Danielsson, A., & Morcuende, J. A. (2008). Adolescent idiopathic scoliosis. The Lancet, 371(9623), 1527-1537.

Weinstein, S. L., Dolan, L. A., Spratt, K. F., Peterson, K. K., Spoonamore, M. J., & Ponseti, I. V. (2003). Health and function of patients with untreated idiopathic scoliosis: A 50-year natural history study. Jama, 289(5), 559-567.

Wettenschwiler, P. D., Lorenzetti, S., Stämpfli, R., Rossi, R. M., Ferguson, S. J., & Annaheim, S. (2015). Mechanical predictors of discomfort during load carriage. PloS one, 10(11), e0142004.

Whittfield, J., Legg, S. J., & Hedderley, D. I. (2005). Schoolbag weight and musculoskeletal symptoms in New Zealand secondary schools. Applied ergonomics, 36(2), 193-198.

Whittfield, J. K., Legg, S. J., & Hedderley, D. I. (2001). The weight and use of schoolbags in New Zealand secondary schools. Ergonomics, 44(9), 819-824.

Willems, J. M., Jull, G. A., & Ng, J. F. (1996). An in vivo study of the primary and coupled rotations of the thoracic spine. Clinical Biomechanics, 11(6), 311-316.

Wilson, R. M., & Gibberd, R. W. (1990). Allocating resources between health care regions: Catchment populations and allocating for equity. European Journal of

Operational Research, 49(3), 333-347.

Winsmann, F. R., & Goldman, R. F. (1976). Methods for evaluation of load-carriage systems. Perceptual and Motor Skills, 43(3 suppl), 1211-1218.

Wu, G., & Cavanagh, P. R. (1995). ISB recommendations for standardization in the reporting of kinematic data. Journal of biomechanics, 28(10), 1257-1261.

Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., ... & Schmid, O. (2002). ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine. Journal of biomechanics, 35(4), 543-548.

Yen, S. C., Gutierrez, G. M., Ling, W., Magill, R., & McDonough, A. (2012). Coordination variability during load carriage walking: can it contribute to low back pain?. Human movement science, 31(5), 1286-1301.

Young, I. A., Haig, A. J., & Yamakawa, K. S. (2006). The association between backpack weight and low back pain in children1. Journal of Back and Musculoskeletal Rehabilitation, 19(1), 25-33.

Yu, Y. Z., & Lu, S. (1990). The acceptable load while marching at a speed of 5km h-1 for young Chinese males. Ergonomics, 33(7), 885-890.

Zeni, J. A., Richards, J. G., & Higginson, J. S. (2008). Two simple methods for determining gait events during treadmill and overground walking using kinematic data. Gait & posture, 27(4), 710-714.

Zhang, X. A., Ye, M., & Wang, C. T. (2010). Effect of unilateral load carriage on postures and gait symmetry in ground reaction force during walking. Computer Methods in Biomechanics and Biomedical Engineering, 13(3), 339-344.

Zheng, N., Fleisig, G. S., Escamilla, R. F., & Barrentine, S. W. (1998). An analytical model of the knee for estimation of internal forces during exercise. Journal of Biomechanics, 31(10), 963-967.

Zultowski, I., & Aruin, A. (2008). Carrying loads and postural sway in standing: the effect of load placement and magnitude. Work, 30(4), 359-368.