Human Movement Science 66 (2019) 164–172

Contents lists available at ScienceDirect

Human Movement Science

journal homepage: www.elsevier.com/locate/humov

Changes of lumbosacral joint compression force profile when walking caused by backpack loads

Simon S.W. Li[a], Yong-Ping Zheng[b], Daniel H.K. Chow[a,*]

[a] Department of Health and Physical Education, The Education University of Hong Kong, Hong Kong
[b] Department of Biomedical Engineering, The Hong Kong Polytechnic University, Hong Kong

ARTICLE INFO

Keywords:
Walking
Backpack
Spine load
Gait
Force profile

ABSTRACT

Walking with backpack loads induces additional mechanical stress on the spine and has been identified as a risk factor of lower-back pain. This study evaluated the effects of walking with backpack loads on the lumbosacral joint compression force profile in both the magnitude and time domains. Ten male adults geared with anatomical markers and trunk surface electromyographic sensors walked along a walkway embedded with three force plates with no load and various backpack loads (5%, 10%, 15%, and 20% body weight). Lower-body movements, ground reaction forces, and trunk muscle activations were measured using a synchronized motion analysis, force plate, and surface electromyography system. The force profiles of identified gait cycles were predicted using an integrated inverse dynamic and electromyography-assisted optimization model and evaluated statistically. The results showed that as backpack load increased, the 10th, 50th, and 90th percentiles of force profiles escalated disproportionately. However, no significant changes were observed in the timing of the two peak force incidences. Such changes in the compression force might be an indication of the combined effects of the increase in both gravitational and mass moment of inertia of the system (body plus pack loads) when walking with a backpack. Pearson correlation coefficients of the force profiles between the five loading conditions were greater than 0.94. Strong associations between the force profiles at different backpack loads were confirmed.

1. Introduction

Walking induces cyclic stress on the spine (Callaghan, Patla, & McGill, 1999), and walking with backpack carriage imposes additional loads in the lumbar spine (Goh, Thambyah, & Bose, 1998). The biomechanical process of walking with backpack loads is extremely complex (Ren, Jones, & Howard, 2005). The increased spinal stress may be caused by the change in pelvis and trunk movement variability (Yen, Gutierrez, Ling, Magill, & McDonough, 2012), non-synchronized coordination between the trunk and backpack (Pierrynowski, Winter, & Norman, 1981), or increases in trunk stiffness (Holt, Wagenaar, LaFiandra, Kubo, & Obusek, 2003), flexion (Singh & Koh, 2009), and ground reaction forces passed through the lower limbs up to the lumbar spine (Birrell, Hooper, & Haslam, 2007). Heavy backpack carriage generates excessive mechanical stress on the lumbar spinal levels and has been considered a risk factor for lower-back problems when engaging in prolonged walking activities (Knapik, Reynolds, Santee, & Friedl, 2012).

Studies on lumbar spinal loads during walking (without carrying a backpack load) have observed that the peak compression force generated on the lower spinal level is between 1.43 and 3.09 (mean: approximately 1.85) times of body weight (BW), and the timing

* Corresponding author at: Department of Health and Physical Education, The Education University of Hong Kong, 10 Lo Ping Road, Tai Po, New Territories, Hong Kong.
E-mail address: danielchow@eduhk.hk (D.H.K. Chow).

https://doi.org/10.1016/j.humov.2019.04.006
Received 4 September 2018; Received in revised form 27 February 2019; Accepted 10 April 2019
Available online 25 April 2019
0167-9457/

of peak force incidence is between the ipsilateral foot initial contact and contralateral foot toe-off (Callaghan et al., 1999; Cheng, Chen, Chen, & Lee, 1998; Goh et al., 1998; Khoo, Goh, & Bose, 1995; Li & Chow, 2018; Shojaei, Hendershot, Wolf, & Bazrgari, 2016). Only two of these studies have evaluated spinal loads, and none have investigated the timing of peak force incidence when walking with backpack loads. Compared with a no-load condition, walking with a backpack load at 10%, 15%, 20%, and 30% BW significantly increases the peak lumbosacral joint compression force by 7%, 23%, 31%, and 64%, respectively, on average (Goh et al., 1998; Li & Chow, 2018). The disproportional increases in lumbar joint loads between carrying heavy and light weight backpack loads demonstrates the significant changes in the loading manoeuver of the lumbar spine (Li & Chow, 2018). However, in-depth and comprehensive study of the lumbar joint force profile under various backpack loads is lacking.

Lumbar joint loads could be simply calculated based on proven biomechanical formula and computerized models by neglecting the input of surface electromyography (EMG) as a measure of the muscle activity around the sacroiliac joint. However, previous studies have validated that optimization-driven model (based on proven biomechanical formula and computerized models) without the measurement inputs of surface EMG values fails to predict substantial antagonistic muscle activities during the performance of dynamic exercises, and thus underestimates the lumbar joint loads (Hajihosseinali, Arjmand, Shirazi-Adl, Farahmand, & Ghiasi, 2014). The novelty of this study is the employment of surface EMG as a measure of trunk muscle coactivation force contributed to the lumbosacral joint compression force during walking exercise. EMG-assisted optimization approach adopted in this study considered both the optimization driven approach accounting for the moment equilibrium at certain spinal level and the surface EMG measurements accounting for the antagonistic muscle activities, and thus predict comparatively accurate muscle forces and lumbosacral joint loads (Arjmand, Gagnon, Plamondon, Shirazi-Adl, & Lariviere, 2009).

This study evaluated the effect of backpack loads on the changes in lumbosacral joint compression force profile during a walking exercise in terms of both time and magnitude. The hypothesis was that incremental increases in backpack load generate quantitatively larger magnitudes and qualitatively comparable profiles of lumbosacral joint compression force during a walking exercise.

2. Methods

2.1. Participants

A convenience sample of ten healthy male young adults (body mass: 71.3 ± 8.9 kg, body height: 1.76 ± 0.04 m, age: 23.7 ± 3.0 y) was recruited from a local university. The number of participants recruited in this study was estimated based on previous similar studies (Goh et al., 1998; Majumdar, Pal, & Majumdar, 2010; Majumdar, Pal, Pramanik, & Majumdar, 2013); it was evaluated using partial Eta-squares (η_p^2), between 0.5 (medium) and 0.8 (large), of individual outcome measures (Cohen, 1992). Participants were screened, and those with known history of any spinal deformity or lower-back pain were excluded. Ethical approval from the Human Research Ethics Committee of XXX and written informed consent from all participants were obtained prior to experimentation.

2.2. Experimental protocol

A repeated-measure design was used to investigate the effects of backpack load (no load: 0% BW; loaded: 5, 10, 15, and 20% BW), and side (left and right foot loading response phase, time between ipsilateral foot initial contact and contralateral foot toe-off) when appropriate, on the spatiotemporal parameters and lumbosacral joint compression force profile in both the time and magnitude domains during a walking exercise. The outcome measures were (i) cadence, walking speed, stride time, and loading response duration; (ii) magnitudes of the force profile; (iii) timing of peak lumbosacral joint compression force incidence in the left and right foot loading response phase; and (iv) lumbosacral joint compression force profiles and correlation coefficients (Cao, Sun, & Zhang, 2017; Wren, Do, Rethlefsen, & Healy, 2006), measuring the similarity of force profiles among the five loading conditions by considering both the time and magnitude domains simultaneously.

An eight-camera motion capturing system (Qualisys, Gothenburg, Sweden) was used to capture the motion of lower-body segments at a sampling rate of 100 Hz. The recoding of 3D motion analysis system is sensitive to reflective objects and electromagnetic fields, thus it is really important to calibrate the system prior to recording so as to improve the reliability of kinematic data. The system was calibrated by a wand with length of 601.2 mm moving randomly across the recorded field for 80 s prior to the data capturing for each experimental section. Calibration was accepted if the average residual was below 0.5 mm for all cameras (Senior, 2004). A surface electromyography (EMG) system (Trigno, Delsys, Boston, MA, USA) was adopted to measure trunk muscle activity at sampling rate of 2000 Hz. Three force plates (Model 4060-10, Bertec Corporation, Columbus, OH, USA) were applied to record the ground reaction forces at sampling rate of 2000 Hz. Qualisys Tracking Manager software (Qualisys, Gothenburg, Sweden) incorporated with a trigger pulse device were set up to synchronize all data during acquisition. A double-strap backpack was used in this study (Li & Chow, 2016).

2.3. Data collection and processing

Twenty anthropometric measurements (malleolus width and height, foot breadth and length, knee diameter, shank circumference and length, mid-thigh circumference, thigh length, anterior superior iliac spine breadth, and body mass) were recorded for estimating the parameters of body segments (Vaughan, Davis, & O'Connor, 1999). Two static anatomical markers (posterior superior iliac spines) and 15 dynamic anatomical markers (sacrum, anterior superior iliac spines, femoral wands, lateral femoral epicondyles, tibial wands, lateral malleoli, heels, and second metatarsal heads) were adhered to their respective anatomical landmarks (Vaughan et al., 1999). The attachment of markers at the wands were to avoid the collinearity of markers for the identification of segmental movements during dynamic activity. Kadaba, Ramakrishnan, and Wootten (1990) of the Helen Hayes Hospital in upstate New York proposed a marker system that uses wands or sticks about 7–10 cm long attached to the thighs and calves. The advantage of this approach is that the markers are easier to track in 3-D space with video-based kinematic systems, and they can (at least theoretically) provide more accurate orientation of the segment in 3-D space. The two static markers attached to the posterior superior iliac spines were used to for calibrating the location of the lumbosacral joint center (Reed, Manary, & Schneider, 1999) while the 15 dynamic anatomical markers attached to the pelvis and lower limbs were used to track the movements of the seven lower body segments during walking exercise. The skin around the 12 identified positions of EMG sensor placement was cleaned by using alcohol to remove dead skin, oil, and dirt as well as shaved if needed to minimize resistance during the acquisition of EMG values representing the muscle activities (McGill, 1992). After skin preparation, 12 EMG electrodes (mass: 14.7 g; length: 37 mm, width: 27 mm, and height: 15 mm; employing four parallel silver bar of 5 × 1 mm contacts perpendicular to the muscle fiber for detecting the EMG signal at the skin surface) were affixed to individual participants' identified surfaces of back and abdominal muscles: the lumbar erector spinae, thoracic erector spinae, latissimus dorsi, internal oblique, external oblique, and rectus abdominis (McGill, 1992). The locations and arrangements of these 6 pairs of muscle groups have been proved to be the best indication of differential muscle activities and to reduce the signal crosstalk generated during twisting and bending tasks (Lafortune, Norman, & McGill, 1988). Participants were asked to maintain a barefoot upright standing posture on the force plates for 3 s for static trial data acquisition, and then instructed to stand at the beginning position of the walkway for 5 s before conducting the walking exercise at the individual's preferred speed along a 10-m walkway (embedded in the middle with three force plates) under the no load (0% BW) and 4 backpack load conditions (5, 10, 15, and 20% BW). This study was intended to observe the changes of the compression force profile when walking (for a short distance) caused purely by backpack loads. Thus, each participant only walked for a short period and distance so as to avoid the probable impact of fatigue occurred during the performance for each experimental section. The level of center of gravity of the backpack under various loaded conditions was standardised at the T12 spinal level by lengthening or shortening shoulder straps (Li & Chow, 2016). Three force platforms (with 2 mm gap between them) were embedded longitudinally in the midway of the 10-m walkway at the level of the concrete floor of the laboratory. The starting position was located for each participant so that he could naturally strike within the lower boundary, middle and upper boundary of the first, second and third force platforms, respectively, for increasing the chance of acquiring the clean strides on three force platforms. The sequence of walking exercises in various loading conditions was randomized. Each participant performed the walking trials repeatedly for various loading conditions until 3 successful trials (clean strides on the 3 force platforms) were acquired for each loading condition. Unsuccessful trials (with unclean strides on the force platforms) were discarded and not used for data processing and analysis. Before the static trial data acquisitions and walking exercises, participants were required to do 4 maximal voluntary contraction (MVC) activities: bent sit up, trunk extension, shoulder extension, and back lift pull (Gagnon, Lariviere, & Loisel, 2001). MVC exercises were recorded for 3 s. Each participant rested for 1–2 min between 2 experimental trials and five min after completing a set of walking exercises or MVC activities. Raw EMG data were full-wave rectified and passed through a bidirectional 4th-order Butterworth low-pass filter at a cutoff frequency of 2.5 Hz (Pakzad, Fung, & Preuss, 2016). The selection of a cut off frequency of 2.5 Hz of the linear envelop was in agreement with the approximation of the tension profile of the muscles representing the activity of the motor unit action potential validated by previous studies (Preuss & Fung, 2008; Winter, 2009). Marker and force platform data were passed through a dual-pass 4th-order Butterworth low-pass filter at cutoff frequencies of 6 and 25 Hz, respectively (Hendershot & Wolf, 2014). All signal processing was performed in MATLAB 2015b software (The MathWorks, Inc., Natick, MA, USA). An identified gait cycle of each walking exercise was observed between two consecutive left foot initial contacts on the force plates (Cappellini, Ivanenko, Poppele, & Lacquaniti, 2006; Zeni, Richards, & Higginson, 2008). The spatiotemporal parameters were acquired in accordance with the gait cycle events: ipsilateral foot initial contact, contralateral foot toe-off, contralateral foot initial contact, ipsilateral foot toe-off, and subsequent ipsilateral foot initial contact (Ayyappa, 1997). Gait cycles of all walking exercise were time-normalized to 101 points for subsequent data analyses. Filtered EMG data of each walking exercise was normalized using the corresponding EMG at MVC. Standardized joint coordinate system (Wu et al., 2002) and kinematic data reporting protocol (Wu & Cavanagh, 1995) were used to establish the anatomical and global coordinate systems. The center of lumbosacral joint was identified based on anatomical landmarks of the sacrum and posterior and anterior superior iliac spines (Reed et al., 1999). A lower-body inverse dynamic approach (comprising seven lower body segments) was used to determine the contact moment and force in the lumbosacral joint (Plamondon, Gagnon, & Desjardins, 1996). The moment acting on the lumbosacral joint was then allocated among the 6 pairs of back and abdominal muscle groups by a musculoskeletal architecture depicting the muscle lines of action, lever arms, and cross-sectional areas (McGill & Norman, 1986; Gagnon et al., 2001). An EMG-assisted optimization approach (Cholewicki, McGill, & Norman, 1995; Gagnon et al., 2001) was used to predict muscle forces. The objective of the optimization processing was to achieve the perfect moment equilibrium in the lumbosacral joint while making the fewest possible adjustments to the initial estimations of muscle force. The lumbosacral joint loads were estimated through summation of the resultant joint contact and trunk muscle forces determined by the inverse dynamic and EMG-assisted optimization approaches, respectively. Details of the computation processes are explained in a previous study (Li & Chow, 2018).

2.4. Statistical analysis

Shapiro–Wilk testing was adopted to verify the normality assumption of all data sets. Two-way or one-way repeated-measure ANOVA was employed to analyze the effect of backpack load and side, when applicable, on the dependent variables: cadence, walking speed, stride time, loading response duration, timing and magnitude of peak force incidence, and force magnitude at the 10th, 50th, and 90th percentile of the force profile. These nonparametric percentile measurements were adopted to represent the lower, medium, and higher amplitude levels of the force profile, respectively. Such measurements of the amplitude probability distribution function have been used to evaluate muscle activities of the neck and shoulder muscles activation (in terms of surface

Table 1
Pooled means, standard deviations, and results of statistical tests of the spatiotemporal parameters of the identified gait cycle for all participants.

Spatialtemporal parameters		Backpack Load (% of body weight)					Overall	One way repeated measure ANOVA	Two way repeated measure ANOVA
		0	5	10	15	20			
Cadence (steps/min)		120 (8.2)	120 (5.7)	119 (6.3)	118 (7.8)	117 (7.6)	119 (7.0)	p = 0.358	–
Walking Speed (m/s)		1.29 (0.11)	1.30 (0.08)	1.27 (0.11)	1.27 (0.09)	1.25 (0.09)	1.28 (0.09)	p = 0.329	–
Stride Time (s)		1.00 (0.07)	1.00 (0.05)	1.02 (0.05)	1.02 (0.07)	1.03 (0.06)	1.01 (0.06)	p = 0.357	–
Loading Response Duration (ms)	Left foot	111 (19)	102 (6)	119 (9)	121 (15)	113 (19)	111 (16)	–	Side (p = 0.080)
	Right foot	101 (10)	110 (14)	108 (23)	111 (8)	114 (28)			Backpack Load (p = 0.072) Interaction (p = 0.137)
Shapiro–Wilk normality test		$p > 0.05$ for all data sets							

EMG measurement) during dynamic activities (Jonsson, 1982; Xie, Szeto, Dai, & Madeleine, 2016) and provide more relevant information regarding the muscle contraction levels of no more than 10% (for the 90th percentile), 50% (for the median or 50th percentile), and 90% (for the 10th percentile) of the dynamic activity involved in completing the exercise. The nonparametric approach may more probably represent the amplitude distribution under various backpack loads during walking exercise.

Pairwise Pearson correlation testing was employed on the force profiles (0%–5%, 0%–10%, 0%–15%, 0%–20%, 5%–10%, 5%–15%, 5%–20%, 10%–15%, 10%–20%, and 15%–20% BW) (SPSS version 21.0, IBM Inc., Chicago, IL, USA). Post hoc tests were performed based on Bonferroni adjustment of multiple comparisons. The significance level was set at $p = 0.05$ for all statistical tests.

3. Results

The pooled means, standard deviations (SDs), and results of statistical tests of the spatiotemporal parameters of the identified gait cycles were determined based on the 150 walking trials (ten participants × five loading conditions × three trials) (Table 1). Shapiro–Wilk tests validated the normality assumption of the data set ($p > 0.05$). One-way repeated-measure ANOVA showed no significant effects of backpack load on cadence ($p = 0.358$), walking speed ($p = 0.329$), or stride time ($p = 0.357$). Two-way repeated-measure ANOVA showed no significant effects of side (left and right foot, $p = 0.080$), backpack load ($p = 0.072$), or interaction ($p = 0.137$) on the loading response duration. The overall mean (SD) of cadence, walking speed, stride time, and loading response duration were 119 (7.0) steps/min, 1.28 (0.09) m/s, 1.01 (0.06) s, and 111 (16) ms, respectively.

The pooled means of the 10th, 50th, and 90th percentile of the lumbosacral joint compression force profiles ($CPF_{10\text{-pct}}$, $CPF_{50\text{-pct}}$, and $CPF_{90\text{-pct}}$) were calculated based on the 30 force profiles (10 participants × 3 trials) for each of the 5 backpack load conditions (Table 2). Shapiro-Wilk tests validated that the normality assumption of data sets could be assumed ($p > 0.05$). One way repeated measure ANOVA showed significant effect of backpack load on $CPF_{10\text{-pct}}$ ($p < 0.001$, $\eta_p^2 = 0.778$), $CPF_{50\text{-pct}}$ ($p < 0.001$, $\eta_p^2 = 0.885$), and $CPF_{90\text{-pct}}$ ($p < 0.001$, $\eta_p^2 = 0.830$). Bonferroni post hoc comparisons showed that all 3 percentiles for backpack load of 15% and 20% BW were significantly higher than no-load condition. However, there were no significant differences in all 3 percentiles between no-load and 5% BW loading conditions. When backpack load increased incrementally from 0% to 20% BW, the $CPF_{10\text{-pct}}$, $CPF_{50\text{-pct}}$,

Table 2
Pooled means (standard deviations) and results of statistical test of the 10th, 50th, and 90th percentiles of peak lumbosacral joint compression force profiles for all participants.

Percentile of force profile (% of body weight)	Backpack Load (% of body weight)					One way repeated measure ANOVA	Post hoc test with Bonferroni adjustment (significant pair: $p < 0.05$)
	0	5	10	15	20		
10th-pct	71 (14)	71 (13)	83 (12)	91 (15)	106 (16)	$p < 0.001$ ($\eta_p^2 = 0.778$)	0 < 10,15,20 5 < 10,15,20 10 < 20
50th-pct	116 (12)	117 (12)	132 (8)	146 (13)	163 (14)	$p < 0.001$ ($\eta_p^2 = 0.885$)	0 < 10,15,20 5 < 10,15,20 10 < 15,20 15 < 20
90th-pct	164 (17)	159 (22)	175 (16)	192 (20)	211 (21)	$p < 0.001$ ($\eta_p^2 = 0.830$)	0 < 15,20 5 < 15,20 10 < 15,20 15 < 20
Shapiro–Wilk normality test						$p > 0.05$ for all data sets	

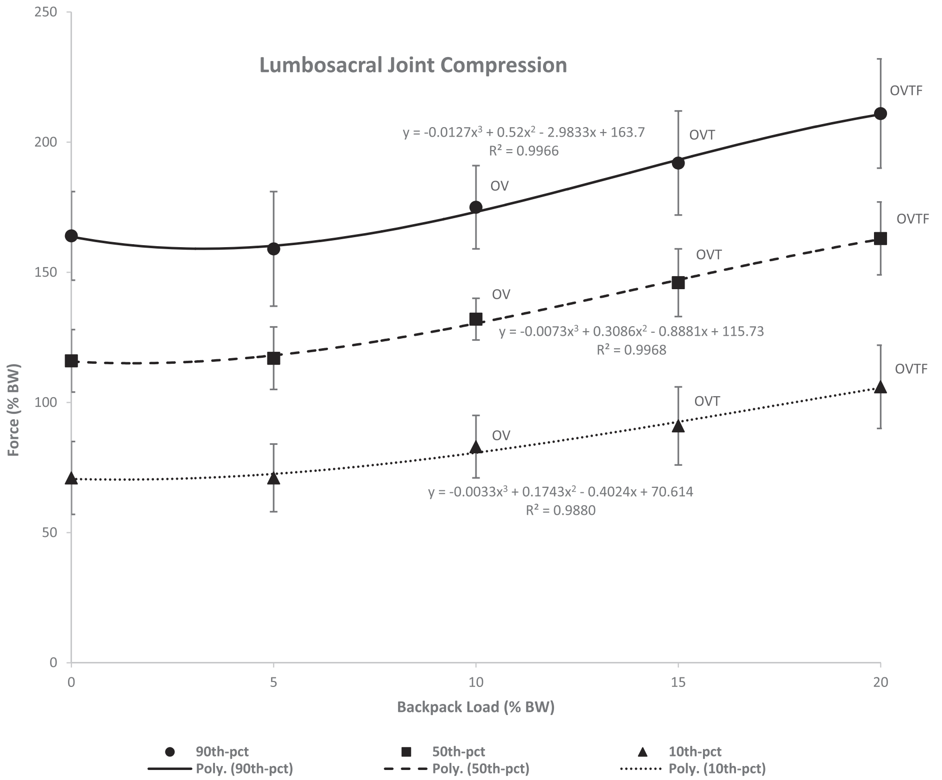

Fig. 1. Regression lines of the pooled mean of the 10th, 50th, and 90th percentile of the lumbosacral joint compression force profiles with respect to the changes in backpack loads of 0%, 5%, 10%, 15%, and 20% BW. Two-sided vertical bars indicate standard deviation of the mean. Data labels 0, V, T, and F indicate significant differences ($p < 0.05$) when comparing with 0%, 5% 10%, and 15%, respectively.

and $CPF_{90\text{-pct}}$ increased incrementally from 71% BW to 106% BW, 116% BW to 163% BW, and 164% BW to 211% BW, respectively.

To evaluate the disproportionately changes in lumbosacral joint compression force magnitude with respect to the changes in backpack loads of 0%, 5%, 10%, 15%, and 20% BW, regression lines were fit to the pooled means of the 10th, 50th, and 90th percentile of the force profiles (Fig. 1). All the three trend lines (third order polynomial regression) were having R-squares greater than 0.98. For all three percentiles, there were no significant difference ($p > 0.05$) in compression force between backpack loads of 0% and 5% BW, and the compression force significantly increased ($p < 0.05$) when backpack load incrementally increased from 5% to 10%, 10% to 15%, and 15% to 20% BW. In general, the lumbosacral joint compression force increased more quickly as backpack load increased from 0% BW through 5%, 10%, 15%, and until reaching the maximum of 20% BW.

The pooled means, SDs, and results of statistical tests of the timing of peak lumbosacral joint compression force incidence were determined based on the 30 force profiles (10 participants × 3 trials) for each of the 5 backpack load conditions (Table 3). Shapiro–Wilk tests validated the normality assumption of the data set ($p > 0.05$). Two-way repeated-measure ANOVA showed no significant effects of side (left and right foot loading response, $p = 0.165$), backpack load ($p = 0.144$), or interaction ($p = 0.315$) on the timing of peak force incidence. The overall mean (SD) of the timing of peak force incidence was 71 (21) ms after the initial contact of the foot on the ground.

To evaluate differential changes in joint contact force (generated by gravitation and inertia) and muscle force (generated by muscle coactivation), the disproportionately changes in the peak lumbosacral joint compression force with various backpack loads were computed and regression lines were fit to the pooled means of the peak contact, muscle and their summation magnitude (Fig. 2). All the three trend lines were having R-squares greater than 0.99. In general, the contact force increased linearly while the muscle force increased non-linearly as backpack load increased from 0% BW through 5%, 10%, 15%, and until reaching the maximum of 20% BW. The contact force, significantly increased ($p < 0.05$) when backpack load incrementally increased from 0% to 5%, 5% to 10%, 10% to 15%, and 15% to 20% BW. However, there were no significant difference ($p > 0.05$) in total and muscle force between backpack loads of 0% and 5% BW, and the two forces significantly increased ($p < 0.05$) when backpack load incrementally increased from 5% to 10%, 10% to 15%, and 15% to 20% BW.

The mean force profiles for each loading condition were obtained (Fig. 3). The Pearson correlation coefficients (in 10 r values) indicating the similarity between two force profiles (0%–5%, 0%–10%, 0%–15%, 0%–20%, 5%–10%, 5%–15%, 5%–20%, 10%–15%, 10%–20%, and 15%–20% BW) were computed (Table 4). All coefficients were greater than 0.94. Strong associations between the force profiles at various backpack loads were validated.

Table 3
Pooled means, standard deviations, and results of statistical tests of the timing of peak lumbosacral joint compression force incidence for all participants.

	Timing after initial contact (ms)									
Side	Left Footˆ					Right footˆ				
Backpack load (% of body weight)	0	5	10	15	20	0	5	10	15	20
Mean	79	68	76	62	65	80	79	78	64	60
SD	26	27	12	18	16	24	22	20	19	18
Overall	Mean: 70 SD: 21					Mean: 72 SD: 22				
	Mean: 71 SD: 21									
Shapiro–Wilk normality test	$p > 0.05$ for all data sets									
Two way repeated measure ANOVA	Side					$p = 0.165$				
	Backpack load					$p = 0.144$				
	Side*backpack load					$p = 0.315$				

ˆ Overall mean (left and right foot) of loading response duration = 111 ms.

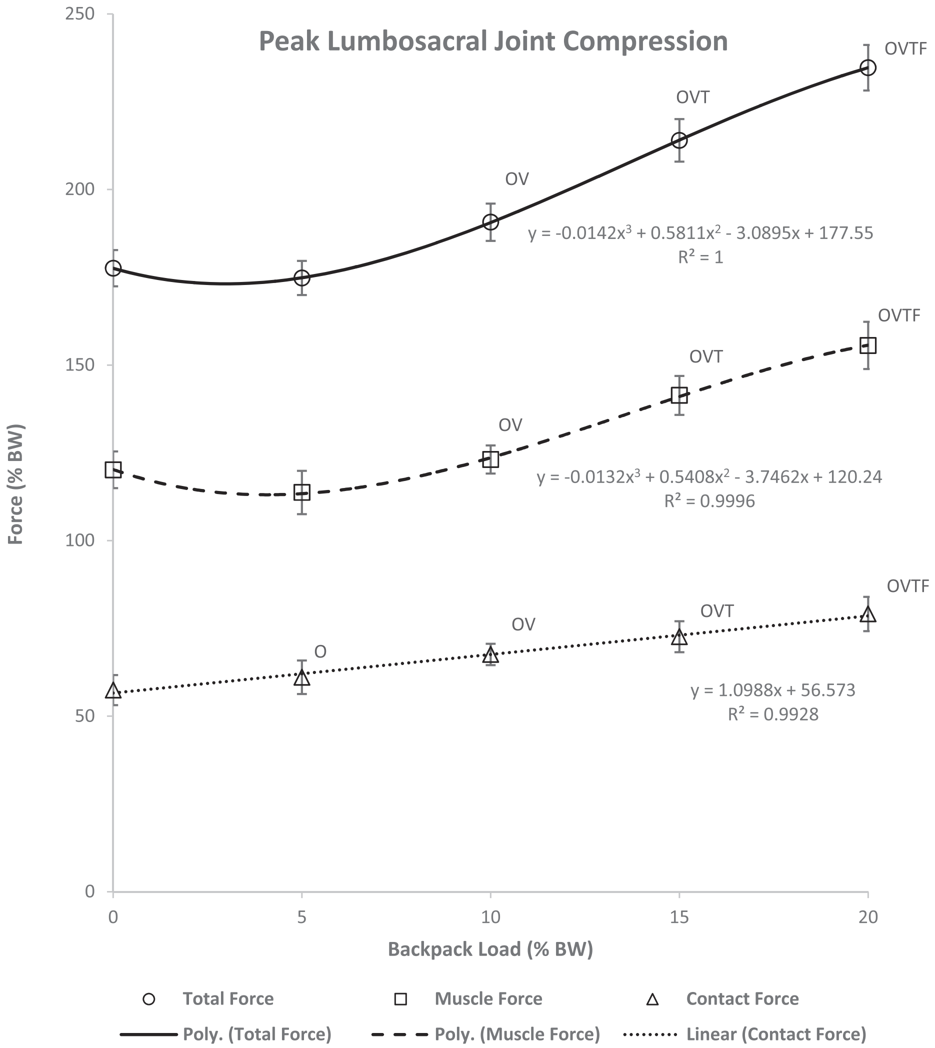

Fig. 2. Regression lines of the pooled mean of the peak lumbosacral joint compression, in terms of the contact (joint contact force generated by gravitation and inertia), muscle (force generated by muscle coactivation), and total (contact plus muscle) force magnitudes with respect to the changes in backpack loads of 0%, 5%, 10%, 15%, and 20% BW. Two-sided vertical bars indicate standard deviation of the mean. Data labels 0, V, T, and F indicate significant differences ($p < 0.05$) when comparing with 0%, 5% 10%, and 15%, respectively.

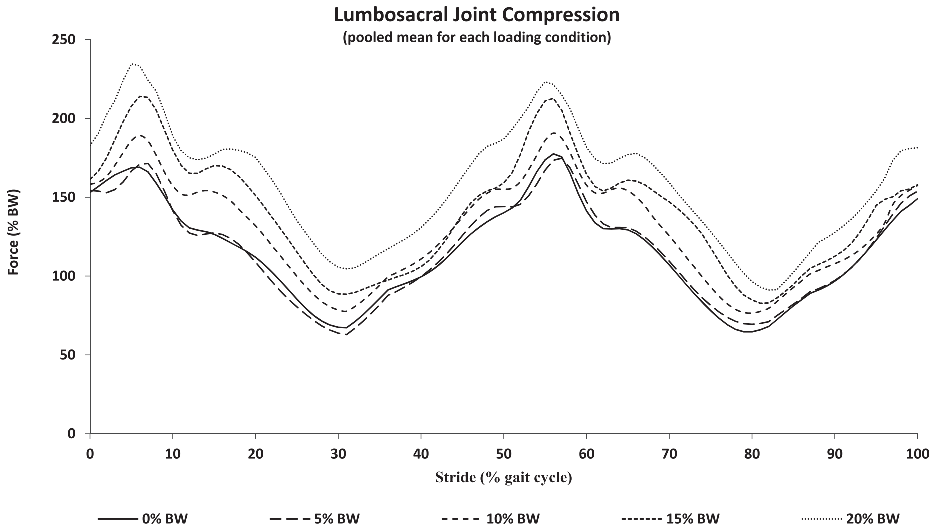

Fig. 3. Profiles of lumbosacral joint compression force for each loading condition. The overall mean of the timing of peak lumbosacral joint compression force incidence was approximately 7% and 57% (equivalent to 71 ms after initial contact of the left or right foot initial contact on the ground) through an identified gait cycle (0% = left foot initial contact, 11% = right foot toe-off, 50% = right foot initial contact, 61% = left foot toe-off, 100% = subsequent left foot initial contact).

4. Discussion

The effects of backpack load on the changes of the lumbosacral joint compression force profile during walking exercise were investigated in both time and magnitude domains. The lumbosacral joint compression force profiles during walking under five repeated-measure conditions (no load and backpack loads of 5%, 10%, 15%, and 20% BW) were compared. Results demonstrated that incremental increases in backpack loads generated quantitatively larger magnitudes and qualitatively comparable lumbosacral joint compression force profiles during a walking exercise.

The observations of this study verified that there were no significant changes of spatiotemporal parameters in terms of cadence, walking speed, stride time, or loading response duration under various loading conditions between 0% and 20% BW. These findings were in agreement with similar research in young adults carrying backpack loads up to 30% BW (Devroey, Jonkers, De Becker, Lenaerts, & Spaepen, 2007; Goh et al., 1998; Liew, Morris, & Netto, 2016; Majumdar et al., 2013).

The effects of increasing backpack loads during the walking exercise became more prominent with the increase of peak lumbosacral joint loads because the spine is required to support more weight (Khoo et al., 1995; Li & Chow, 2018). Such increase in joint loads results from the combined kinematic (postural changes in both the upper and lower body) and kinetic (changes in muscle coactivity) effects of the dynamic control required to maintain stability of the upper body and forward progression during walking (Devroey et al., 2007; Khoo et al., 1995). Notably, the increase in magnitude of the percentiles of lumbosacral compression forces was not in same proportion as the increase in the backpack loads between 0% and 20% BW. The increment of the compression force from 0% to 5% BW was minimal (only 1% BW) for the 10th and 50th percentiles; a reduction of 5% BW was observed for the 90th percentile. Such increments (even the reduction for 90th percentile) were comparably less than those increments (8%–19% BW) between backpack loads of 5%–10%, 10–15%, and 15–20% BW. The disproportional increments were similar to the findings in peak force at backpack loads of 0%–5%, 5%–10%, 10%–15%, 15–20%, and 15%–30% BW (Goh et al., 1998; Li & Chow, 2018). Three studies reported that carrying backpack loads at 10% BW or above significantly changes trunk muscle activation levels, spine kinematics, lumbopelvic coordination, and lumbar spine load-bearing strategy (Chow, Wang, & Pope, 2014; Devroey et al., 2007; Li & Chow, 2018). Such kinematic and kinetic responses with respect to lighter weight ($<$ 10% BW) and heavier weight ($\geq$10% BW) might cause disproportionate changes in lumbosacral compression force and indicate the beneficial combined effect of increase in both gravitational and mass moment of inertia of the system (body plus backpack) during walking with a light weight backpack of 5%

Table 4
Pearson correlation coefficients of pairwise comparisons between lumbosacral joint compression force profiles at various backpack load.

Backpack load (% body weight)	0	5	10	15	20
0	–	–	–	–	–
5	0.993*	–	–	–	–
10	0.986*	0.983*	–	–	–
15	0.950*	0.942*	0.971*	–	–
20	0.963*	0.953*	0.978*	0.986*	–

* Indicates $p < 0.001$.

BW (Costello, Matrangola, & Madigan, 2012; Li & Chow, 2018).

The timing of two peak force incidences appeared after initial contact of the ipsilateral foot on the ground and before the contralateral foot toe-off. The timing of two minimum force incidences occurred during the single-leg supporting phase (approximately 30% and 80% of the gait cycle). The results are comparable with previous studies (Callaghan et al., 1999; Cheng et al., 1998; Goh et al., 1998). When backpack load increases, the weight of the moving system (body plus loaded backpack) increases. To generate a larger progressive forward force, the neuromuscular system caters an adaptive response in terms of the increases in hip and knee extensor moments to provide shock absorption and support more weight during the loading response phase (Wang, Frame, Ozimek, Leib, & Dugan, 2013). However, such response is not apparent when backpack loads are less than 40% BW (Wang et al., 2013). It is likely that the observed invariance in the timing of peak force incidence occurred in the later stage of the loading response phase (71 ms after foot initial contact) to generate more power at the ankle joint (Huang & Kuo, 2014) and is the result of the combined effects of nonsignificant changes in both lower-limb kinetic responses and spatiotemporal parameters.

The similarity of lumbosacral compression force profiles among various loading conditions confirmed that the main difference was the magnitude of the profile, not the timing of the peak or minimum force incidences, during the walking exercise. As backpack load increased, the contact force increased linearly while the muscle force increased non-linearly (Fig. 2). As backpack loads increased, the centre of mass of the system (BW plus backpack load) shifted posteriorly. To adopt such kinematic and kinetic changes, the trunk flexed forward and shifted the center of mass anteriorly for the spine to support more weight. Such postural change generated additional mass moment of inertia to stabilize the body during the walking exercise. The load-bearing manoeuver of the lower-limb movement was verified by the nonsignificant changes in the spatiotemporal parameters, including cadence, walking speed, stride time, and loading response duration under various loading conditions. The strong association of the force profile demonstrated similarity in the time domain rather than the magnitude of the compression force generated on the lumbar spine. Moreover, this study observed that the exponentially increase in lumbosacral joint compression force with backpack loads was due to the differential effects of the resultant forces generated by the gravitation and inertia as well as the muscular co-activation.

This study has serval limitations. First, for the convenience of conducting the experiment, all participants recruited in this study were young adult men; thus, the findings may not be generalizable to other genders or age groups. Second, this study investigated the effects of short-term backpack load carriage; thus, the observations may not be applicable to long-term carriage, because trunk muscle activity may vary with fatigue and have significant effects on lumbar joint loads. Third, only five backpack loading conditions and a small sample size of ten participants were evaluated, which might be a concern of generalizing the conclusions of this study. More detailed and larger scale research is recommended for future study on the effect of backpack loads on lumbar spine loading or at the joints of the lower limbs during walking. Fourth, this study evaluated the magnitude and time domain of lumbosacral joint compression force profile for a walk of 10 m only with an average duration of 7.8 s. The results reported no significant time difference among the five backpack load conditions, which might be due to the short walking period and distance, and thus the findings of this study may not be applicable for longer walk. Further study is recommended to evaluate the effects of backpack loads on the time difference after longer walk of 10–15 min. Fifth, a previous study has demonstrated a significant effect of walking speed on lumbosacral joint loads: higher walking speed generates higher lumbar disc compression (Cheng et al., 1998). The different preferred walking speeds adopted by each participant from various loads observed in this study might be a potential confounding effect on the changes in lumbosacral joint loads.

5. Conclusion

No significant changes were observed in the timing of two peak force incidences appearing after the initial contact of the ipsilateral foot on the ground and before the contralateral foot toe-off. As backpack load increased, the 10th, 50th, and 90th percentiles of force profiles escalated disproportionately. Such changes in lumbosacral compression force might indicate the combined effect of increase in both gravitational and mass moment of inertia of the system (body plus backpack) when walking with a backpack. Moreover, Pearson correlation coefficients of the force profiles among the five loading conditions were greater than 0.94. Strong associations among the force profiles at various backpack loads were confirmed.

Conflict of interest disclosure

None.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.humov.2019.04.006.

References

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.

Ayyappa, E. (1997). Normal human locomotion, Part 1: Basic concepts and terminology. *JPO: Journal of Prosthetics and Orthotics, 9*(1), 10–17.

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.

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.

Cao, H., Sun, S., & Zhang, K. (2017). Modified EMG-based handgrip force prediction using extreme learning machine. *Soft Computing, 21*(2), 491–500.
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.
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.
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., 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.
Cohen, J. (1992). A power primer. *Psychological Bulletin, 112*(1), 155.
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.
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.
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.
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.
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 & Physics, 36*(10), 1296–1304.
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.
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.
Huang, T. W., & Kuo, A. D. (2014). Mechanics and energetics of load carriage during human walking. *Journal of Experimental Biology, 217*(4), 605–613.
Jonsson, B. (1982). Measurement and evaluation of local muscular strain in the shoulder during constrained work. *Journal of Human Ergology, 11*(1), 73–88.
Kadaba, M. P., Ramakrishnan, H. K., & Wootten, M. E. (1990). Measurement of lower extremity kinematics during level walking. *Journal of Orthopaedic Research, 8*(3), 383–392.
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.
Knapik, J., Reynolds, K., Santee, W. R., & Friedl, K. E. (2012). Load carriage in military operations: A review of historical, physiological, biomechanical and medical aspects. *Military Quantitative Physiology: Problems and Concepts in Military Operational Medicine* (pp. 303–337). Ft Detrick: Office of the Surgeon General and the Borden Institute.
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.
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.
Liew, B., Morris, S., & Netto, K. (2016). The effect of backpack carriage on the biomechanics of walking: A systematic review and preliminary meta-analysis. *Journal of Applied Biomechanics, 32*(6), 614–629.
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.
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., & Norman, R. W. (1986). Partitioning of the L4–L5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine, 11*(7), 666–678.
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.
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.
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.
Preuss, R., & Fung, J. (2008). Musculature and biomechanics of the trunk in the maintenance of upright posture. *Journal of Electromyography and Kinesiology, 18*(5), 815–828.
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.
Ren, L., Jones, R. K., & Howard, D. (2005). Dynamic analysis of load carriage biomechanics during level walking. *Journal of Biomechanics, 38*(4), 853–863.
Senior, D. (2004). Qualisys Track Manager: User Manual.
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.
Singh, T., & Koh, M. (2009). Effects of backpack load position on spatiotemporal parameters and trunk forward lean. *Gait & Posture, 29*(1), 49–53.
Wang, H., Frame, J., Ozimek, E., Leib, D., & Dugan, E. L. (2013). The effects of load carriage and muscle fatigue on lower-extremity joint mechanics. *Research Quarterly for Exercise and Sport, 84*(3), 305–312.
Vaughan, C., Davis, B., & O'Connor, J. C. (1999). *Dynamics of human gait* (2nd ed.). Cape Town, South Africa: Kiboho Publishers17–23.
Winter, D. A. (2009). *Biomechanics and motor control of human movement* (4th ed.). John Wiley & Sons269–272.
Wren, T. A., Do, K. P., Rethlefsen, S. A., & Healy, B. (2006). Cross-correlation as a method for comparing dynamic electromyography signals during gait. *Journal of Biomechanics, 39*(14), 2714–2718.
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.
Xie, Y., Szeto, G. P., Dai, J., & Madeleine, P. (2016). A comparison of muscle activity in using touchscreen smartphone among young people with and without chronic neck–shoulder pain. *Ergonomics, 59*(1), 61–72.
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.
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.