Journal of Biomechanics 94 (2019) 107–114

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Effects of age and obesity on trunk kinetics and kinematics during dominant side one-handed carrying

Mohamed Badawy [a], Mark C. Schall Jr [a,*], Michael E. Zabala [b], Jordan Coker [b], Richard F. Sesek [a], Sean Gallagher [a], Gerard A. Davis [a]

[a] Department of Industrial and Systems Engineering, Auburn University, Auburn, AL, USA
[b] Department of Mechanical Engineering, Auburn University, Auburn, AL, USA

ARTICLE INFO

Article history:
Accepted 15 July 2019

Keywords:
Dominant side one-handed carrying
Aging
Obesity
Trunk
Kinematics
Kinetics

ABSTRACT

The proportions of older and obese people are increasing in both the general and working populations worldwide. Older and obese individuals are more susceptible to work-related musculoskeletal disorders (MSDs) in comparison with healthy, younger individuals. Manual material handling (MMH) is associated with the development of work-related MSDs. Although previous research has suggested that one-handed carrying is a particularly undesirable method of MMH, the effects of one-handed carrying on trunk kinetics and kinematics among older and/or obese people have not been adequately studied. The objective of this study was to examine the effects of age and obesity on trunk angles and moments during dominant side one-handed carrying of various load magnitudes. Twenty (20) participants divided into four groups with respect to age (young and older) and obesity (obese and non-obese) carried different loads (*No-load* [0 kg], *Light* [5.67 kg], and *Heavy* [10.21 kg]) in their dominant hand for approximately 6 m. Three-dimensional (3D) trunk angles and moments approximately about the L4/L5 vertebral segment were calculated using Visual3D. The findings indicated that while carrying a load in the dominant hand plays an important role in changing trunk kinematics and kinetics, the results were not dependent on age and/or obesity category. Absolute moments were greatest among participants in the obese groups; however, these moments were mitigated when normalized to body weight and height (%BW * Ht). Age did not exacerbate the effects of load magnitude on trunk kinetics and kinematics.

1. Introduction

Approximately 12.3% of the world's population was aged 60+ years in 2015 and the percentage is expected to increase 56% by 2030 and 130% by 2050 (UNDESA, 2015a; UNDESA, 2015b). In the United States, the proportion of the population aged 65+ years is expected to reach approximately 20.9% in 2050; double the percentage reported in 2012 (Ortman et al., 2014). The percentage of older people comprising the American working population is expected to increase as well (Toossi, 2015).

Obesity rates are also increasing worldwide. The global prevalence of overweight and obese adults increased 27.5% between 1980 and 2013 (Ng et al., 2014) while obesity increased 9.1% among American adults between 1999 and 2000 and 2015–2016 (Hales et al., 2017). In 2030, over 50% of American adults are expected to be obese (Wang et al., 2008; Finkelstein et al., 2012). On average, obesity is more prevalent among people aged 45+ years than among younger individuals worldwide (Ng et al., 2014; Hales et al., 2017). A high prevalence of obesity has also been observed among the American working population (Gu et al., 2014; Luckhaupt et al., 2014).

Work-related musculoskeletal disorders (MSDs) are more prevalent among older and obese people in comparison with healthy, younger populations (Kouvonen et al., 2013; BLS, 2015; Schulte et al., 2007). For example, low back pain has been associated with increasing age (Hoy et al., 2012; Wong et al., 2017), and obese people appear to be at higher risk of low back pain in comparison to non-obese people (Sheng et al., 2017; Roffey et al., 2013; Peng et al., 2018). Low back pain risk may be exacerbated for individuals who are both older and obese as they tend to have lower muscle strength than older, non-obese people (Villareal et al., 2004). Accordingly, the changing composition of the working population evokes the necessity for a deeper understanding of work-related health outcomes among these populations.

* Corresponding author at: 3301F Shelby Engineering Center, Auburn, AL 36849, USA.

https://doi.org/10.1016/j.jbiomech.2019.07.016
0021-9290/

Previous research has indicated a strong association between manual material handling (MMH) and the development of work-related MSDs (Xu et al., 2013; Nelson et al., 2006; Gallagher et al., 2017). One-handed carrying has been identified as a particularly undesirable method of MMH since it is more physiologically and biomechanically demanding than other methods of load carrying (e.g., two-handed, anterior, posterior; Cook and Newmann, 1987; McGill et al., 2013; Lind and McNicol, 1968; Ganguli and Datta, 1977). The effects of one-handed carrying on spinal loading has been the focus of two studies (McGill et al., 2013; Rohlmann et al., 2014). McGill et al. (2013) reported larger compression and shear forces on the L4/L5 vertebral segment when young, healthy male participants (aged 22.7 ± 2.1 years) carried a load in one hand relative to when the load was evenly split between two hands, or even when carrying the same load in both hands (twice the carried load). Rohlmann et al. (2014) studied the effects of different load carrying methods (in one hand, in two hands, and on the back) among older individuals (aged 66.2 ± 3.8 years) with implanted (instrumented) vertebral body replacements. Loading on the implants was higher when 10 kg was carried in one hand than when 20 kg were carried in both hands (10 kg in each hand). However, trunk kinematics and kinetics, which play an important role in spinal loading during MMH and in the development of low back disorders (Lee and Nussbaum, 2013; Marras, 2000), have not been studied for one-handed carrying.

The objective of this study was to examine changes in trunk angles and moments about the approximate location of the L4/L5 vertebral segment among older and obese males during dominant side one-handed carrying of various load magnitudes. We hypothesized that obese and older individuals would experience larger changes in trunk kinematics and kinetics in comparison to non-obese and young individuals. This hypothesis was based on previous research suggesting that excess weight of obese people can be disadvantageous during MMH, in comparison with non-obese participants, as it may lead to undesired effects on the musculoskeletal structures of the back and the capacity of bringing the load closer to the body (Corbeil et al., 2019). Obesity is also associated with balance impairments and abnormal gait patterns (Del Porto et al., 2012; Bergland et al., 2000; Guelich, 1999; Zecevic et al., 2006). In addition, Shojaei et al. (2016) suggested that larger pelvic rotation, smaller lumbar flexion, and larger low back peak shearing demand were observed among older participants, in comparison with younger participants, during symmetric MMH tasks. To the best of our knowledge, the current study is the first to evaluate the effects of age and obesity on trunk kinematics and kinetics during one-handed carrying (Badawy et al., 2018). This work provides insights that may be helpful for establishing one-handed carrying guidelines for the rapidly changing working population.

2. Method

2.1. Participants

Twenty (20) right-handed (self-reported), male participants were recruited for this study. Five eligibility criteria were considered: (1) Age of 19–35 or 55–64 years; (2) A body mass index (BMI) < 25 kg/m^2 or BMI ≥ 30 kg/m^2 (WHO, 1995); (3) No history of physician-diagnosed cardiovascular diseases or MSDs in the neck, shoulder, extremities, or low back regions; (4) No chronic pain in the neck, shoulder, extremities, or low back in the 6 months preceding the study; and (5) No adhesive allergies. Eligible participants were divided into four groups (five participants in each group) with respect to age and obesity (Young/Non-obese [YNO], Young/Obese [YO], Older/Non-obese [ONO], and Older/Obese [OO]). After providing their informed consent, all participants had a Dual-energy X-ray absorptiometry (DEXA) scan to estimate their percentage of body fat. All study procedures were approved by the Auburn University Institutional Review Board. Means and standard deviations of age, weight, height, and BMI, %Fat, and fat distribution of each group are presented in Table 1.

2.2. Pre-experiment protocol

This study was performed in the Auburn University Biomechanical Engineering (AUBE) Laboratory. Seventy-nine reflective markers were placed on participants using hypoallergenic tapes. Marker locations are listed in Table 2. A previously validated reflective marker-based point cluster technique was used to collect lower-extremity kinematic data (Andriacchi et al., 1998; Andriacchi and Dyrby, 2005; Dyrby and Andriacchi, 2004). The marker set used for the trunk was similar to the obesity-specific marker set developed by Lerner et al. (2014). Marker placement was completed by one researcher and verified by a second researcher to ensure that anatomical landmarks were properly located. Kinematic data were collected at a sampling rate of 120 Hz using a 10-camera motion capture system (Vicon, Vantage V5 Wide Optics cameras, each with 22 high-powered IR LED strobe at 85 nm; Vicon Motion Systems Ltd, Oxford Industrial Park, Oxford, UK), and ground reaction force data were collected at a sampling rate of 2000 Hz using two force plates (AMTI BP400600, 2000 lb. capacity; Advanced Mechanical Technology, Inc., Watertown, MA).

2.3. Experiment

Participants were asked to perform three carrying conditions, each replicated three times (a total of nine trials). These conditions included walking across the lab (approximately 6 m) while carrying different loads in the dominant/preferred (right) hand (0 kg [*No-load*], 5.67 kg [*Light*], and 10.21 kg [*Heavy*]). The order of the nine walking trials was randomized. The trials were separated by a resting period of at least 1 min to avoid fatigue and/or muscle soreness. Participants were instructed on the starting location(s) of each trial such that each foot contacted only one force plate, while maintaining the normal gait pattern. The two force plates were placed tangentially; no space separated the two force plates. Participants performed rehearsals before data collection started for a research team member to identify the starting line for each participant for each walking/carrying condition to avoid having a participant fix their cadence and/or step length/width to ensure

Table 1
Mean and standard deviation of personal data of each group of participants.

Group	n	Age (years)	Weight (kg)	Height (cm)	BMI (kg/m^2)	%Body fat	Fat mass - Trunk (kg)	Fat mass - Arms (kg)	Fat mass - Legs (kg)
YNO	5	25.4 ± 2.1	70.7 ± 5.3	173.2 ± 5.8	23.7 ± 0.7	15.2 ± 4.2	7.6 ± 2.5	1.6 ± 0.5	5.1 ± 1.5
YO	5	29.1 ± 4.5	113.0 ± 19.4	179.3 ± 4.6	35.0 ± 4.7	39.4 ± 7.0	22.8 ± 3.1	4.3 ± 1.4	11.1 ± 2.5
ONO	5	59.7 ± 3.5	70.2 ± 6.6	175.3 ± 4.4	22.9 ± 0.9	16.0 ± 5.6	8.9 ± 4.1	1.6 ± 0.5	4.6 ± 0.9
OO	5	60.1 ± 0.8	104.2 ± 4.6	180.3 ± 3.6	31.9 ± 1.4	34.0 ± 4.8	21.0 ± 3.3	3.4 ± 0.5	8.4 ± 1.5

YNO: 19–35 years of age, BMI < 25 kg/m^2.
YO: 19–35 years of age, BMI ≥ 30 kg/m^2.
ONO: 55–64 years of age, BMI < 25 kg/m^2.
OO: 55–64 years of age, BMI ≥ 30 kg/m^2.

Table 2
Marker locations (number of markers) used to define a model for each participant.

Head	Trunk	Upper limbs	Lower limbs
Forehead (2)	Sternoclavicular notch (1)	Medial and lateral epicondyles (4)	Greater trochanters (2)
	Xyphoid process (1)	Radius and ulnar styloid processes (4)	Thighs (18)
	Acromion processes (2)	Upper arms (2)	Medial and lateral femoral condyles (4)
	7th cervical vertebrae (1)	Forearms (2)	Medial and lateral tibial plateaus (4)
	Right scapular inferior angle (1)	Dorsal hands (2)	Shanks (12)
	10th thoracic vertebrae (1)		Medial and lateral malleoli (4)
	Iliac crests (2)		1st metatarsal heads (2)
	Anterior–superior iliac spines (2)		5th metatarsal heads (2)
	Posterior–superior iliac spines (2)		Calcanei (2)

contact with each force plate. The loads were handed to each participant by a researcher at their hand height while the arms were at their neutral posture. Participants did not have to bend over (or lean forward or sideways) to pick up or drop the load. Participants waited approximately 2–3 s to stabilize themselves and the load in their hand before they started walking.

2.4. Data processing

A skeletal model of each participant, scaled to their weight and height, was constructed using Visual3D (C-Motion, Germantown, MD, USA). A pipeline was constructed to calculate the kinematic and kinetic results of the trunk along the gait cycle. Joint angles (flexion/extension, ipsilateral/contralateral obliquity, and ipsilateral/contralateral rotation) were calculated using a Cardan sequence of rotations according to the Visual3D default segment coordinate system (z-up, y-anterior; Robertson et al., 2013), consistent with the Joint Coordinate System provided by Grood and Suntay (1983). Body segments were described according to the International Society of Biomechanics recommendations (Wu et al., 2002). The shoulder markers (placed on the acromion) were used to create the distal joints of the trunk segment and the two iliac crest markers were used to create the proximal joints. Markers on the clavicle, sternum, right scapula, and tenth vertebral body were also tracked. Using these markers, a three-dimensional (3D) coordinate system was created for the trunk segment. The 3D angles between the trunk coordinate system and the lab coordinate system were considered the trunk angles.

To calculate reaction moments, a bilateral Newton-Euler model that includes the lumbar and thoracic segments was used (Seay et al., 2008, Robertson et al., 2013). A joint was created between the pelvis and the trunk which represents the waist for all angle and moment calculations (Seay et al., 2008). The pelvis was identified by six markers placed on the left and right iliac crest, anterior superior iliac spine, and posterior superior iliac spine. Based on palpation of the lower back of participants, Chakraverty et al. (2007) suggested that the iliac crest is in line with L3 or the L3/L4 joint. However, based on fluoroscopic imaging, they suggested that the line passing through the two iliac crest markers were found to pass through the L4 or L4/L5 joint. Accordingly, it is assumed that the waist moments in the current study are calculated approximately about the L4/L5 spinal level (Robertson et al., 2013).

Trunk angles, and moments about the L4/L5 vertebral segment, during the stance phase of each limb (right [loaded side] and left [unloaded side]) were calculated. Data of one step during an established gait (after participants walked for 3 steps) was considered. The stance phase of each limb was identified at a ground reaction force of 20 N (Tirosh and Sparrow, 2003). Data during each stance phase (loaded and unloaded) were normalized to 101 time points ($t = 0\%$ to 100%). Angles and moments of the three trials for each carrying condition (*No-load*, *Light*, and *Heavy*) were averaged. The averages of resultant values for each participant group were calculated. Two of the three *Heavy* load trials for one participant of the ONO group were unusable due to data collection error, and hence excluded. Data resulting from the remaining trial was used to represent the *Heavy* load condition for this particular participant.

2.5. Independent and dependent variables

Three independent variables were examined: age (young/older), obesity (obese/non-obese), and load magnitude (*No-load*, *Light*, and *Heavy*). Eighteen dependent variables were analyzed for each stance limb (thirty-six measurements in total). These included 3D trunk angles (tilt [extension/flexion], obliquity [ipsilateral/contralateral], and rotation [ipsilateral/contralateral]) and 3D reaction moments (both absolute and normalized to body weight and height [%BW * Ht]) at the L4/L5 vertebral segment (tilt [extension/flexion], obliquity [ipsilateral/contralateral], and rotation [ipsilateral/contralateral]). Normalized moments were analyzed to identify the effects of the independent variables regardless of differences in weight between groups (Blazek et al., 2013). A description of the 3D motions and sign conventions is presented in Table 3.

2.6. Statistical analysis

Shapiro Wilk and Levene's tests were used to evaluate analysis of variance (ANOVA) assumptions (normality and equality of variance, respectively) using Statistix 8.0 (Analytical Software; Tallahassee, FL, Maryland, USA) and SPSS (SPSS Inc, release 14.0, Chicago, IL) statistics software. Fifteen of 36 measurements passed both Shapiro Wilk and Levene's tests without transformation. These measurements were: (1) Unloaded side: (*angles*: flexion, extension, ipsilateral rotation, contralateral rotation; *moments*: ipsilateral obliquity; *normalized moments*: flexion, ipsilateral obliquity, contralateral obliquity, contralateral rotation), and (2) Loaded side: (*angles*: extension, ipsilateral obliquity, ipsilateral rotation, contralateral rotation; *normalized moments*: flexion, ipsilateral rotation). Transformations were applied as necessary for the remaining measurements. All measurements met the normality assumption following transformations, whereas one measurement (ipsilateral obliquity moment [loaded side]) failed to meet the assumption for equality of variance following transformations. The original (untransformed) data for that specific measurement was used since three-way ANOVA is generally robust to violations of assumptions. The statistical significance of the main and interaction effects of the independent variables on the dependent

Table 3
Descriptions of 3D motions and sign conventions of the trunk.

Motion	Description	Sign
Extension	Leaning toward the back	+
Flexion	Leaning toward the front	–
Ipsilateral obliquity	Side leaning toward the load	+
Contralateral obliquity	Side leaning away from the load	–
Contralateral rotation	Rotating the front of the body away from the load	+
Ipsilateral rotation	Rotating the front of the body toward the load	–

variables were tested using ANOVA following a split-split-plot factorial design (using Statistix 8.0). Tukey honest significant difference (HSD) post-hoc tests were conducted (using Statistix 8.0) to compare each pair of the 3-level load conditions when significant using an alpha value of 0.05, and results are presented in Figs. 1–3. Only statistically significant effects are discussed. Main effects were not discussed if the interaction effect was significant.

3. Results

3.1. Peak trunk angles

3.1.1. Loaded side (stance phase)

Load carrying resulted in a statistically significant change of the 3D trunk angles. All participant groups tended to have a smaller extension angle ($p < 0.01$), ipsilateral obliquity angle ($p < 0.01$), and contralateral rotation angle ($p < 0.01$) during load carrying (Fig. 1). In general, the participants walked with more trunk flexion, more side bending away from the load, and more trunk rotation toward the load relative to the *No-load* condition.

3.1.2. Unloaded side (stance phase)

Results for the unloaded side were generally consistent with those of the loaded side. The only exception was a statistically significant effect of age and load on ipsilateral obliquity angle ($p = 0.04$). The young groups experienced smaller ipsilateral obliquity angles while carrying the *Heavy* load compared to carrying the *Light* load, whereas the older group experienced larger angles.

3.2. Absolute peak L4/L5 moments

3.2.1. Loaded side (stance phase)

A statistically significant increase in extension moment was observed ($p = 0.04$). Carrying the *Heavy* load resulted in an

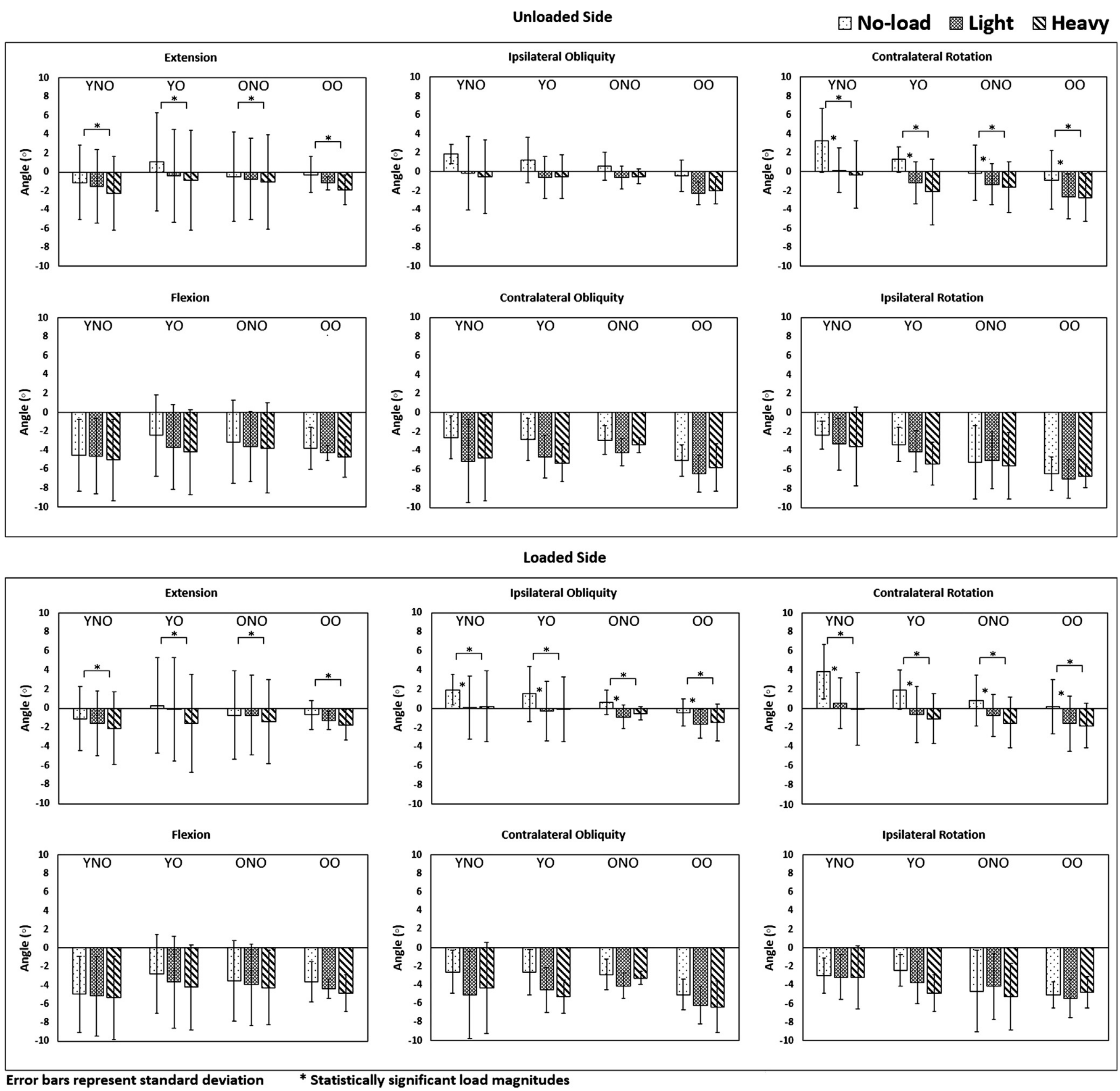

Fig. 1. Peak trunk angles at three load conditions. YNO: Young/Non-obese; YO: Young/Obese; ONO: Older/Non-obese; OO: Older/Obese.

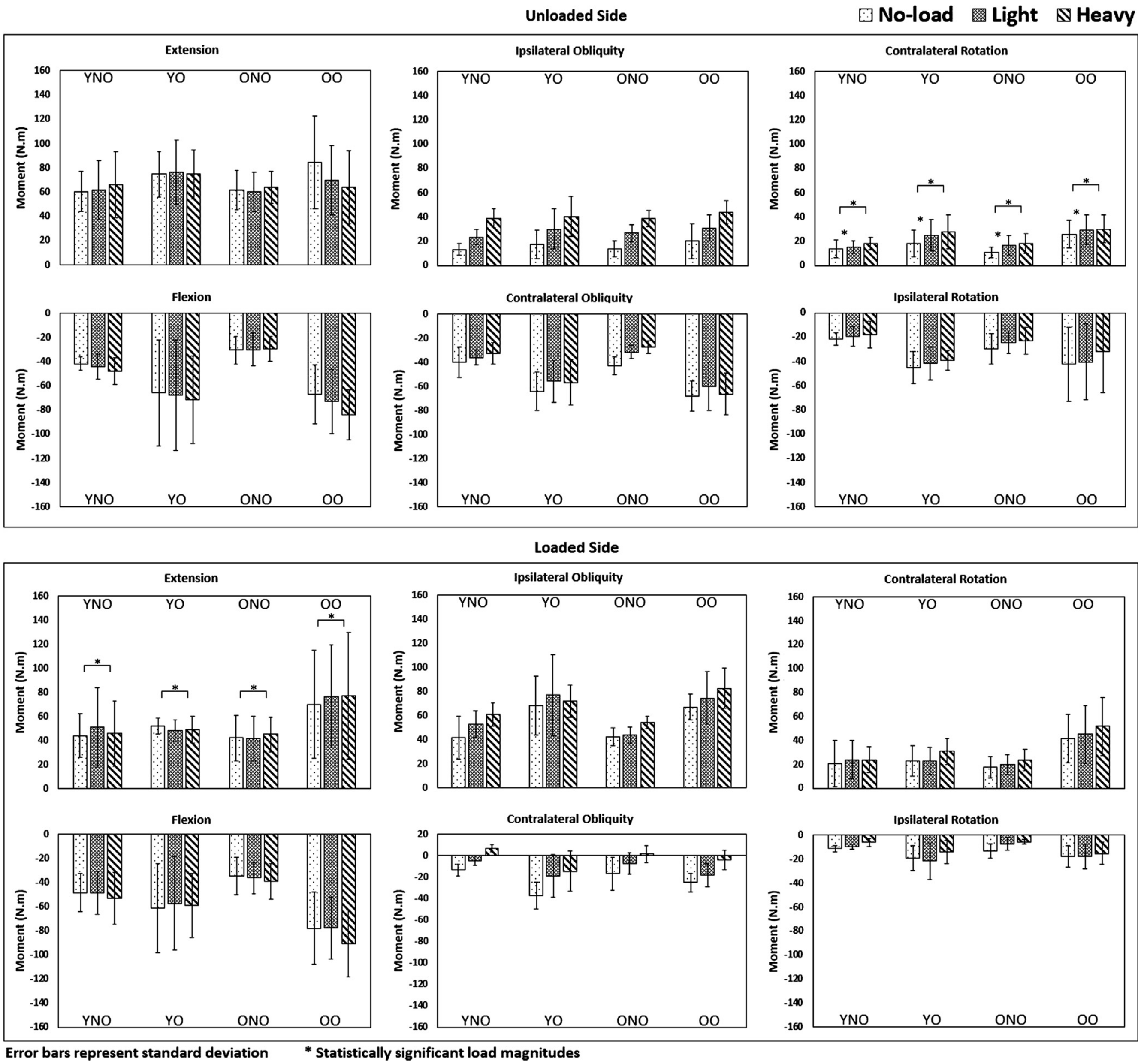

Fig. 2. Absolute peak L4/L5 moments at three load conditions. YNO: Young/Non-obese; YO: Young/Obese; ONO: Older/Non-obese; OO: Older/Obese.

increased extension moment compared with the corresponding moment that resulted from walking with *No-load* for all participant groups except for the YO group (Fig. 2).

3.2.2. Unloaded side (stance phase)

Carrying either the *Light* or the *Heavy* loads resulted in an increased contralateral rotation moment compared with the *No-load* condition ($p < 0.01$). Moments were greater for the obese groups than for the non-obese groups for all three load conditions ($p < 0.02$). Furthermore, a statistically significant interaction between obesity and load was observed for the ipsilateral obliquity moment ($p = 0.02$); moments among the obese groups were more sensitive to change while the *Heavy* load was carried.

3.3. Normalized peak L4/L5 moments

3.3.1. Loaded side (stance phase)

Consistent with the results for the absolute moments, the change in trunk angles due to load carrying resulted in increased flexion moment (except for the YO group; $p = 0.04$), increased ipsilateral obliquity moment ($p < 0.01$), decreased contralateral obliquity moment ($p < 0.01$), and increased contralateral rotation moment ($p < 0.01$), when compared to walking with *No-load* (Fig. 3). A statistically significant interaction between obesity and load was observed on the ipsilateral rotation moment ($p < 0.01$) where the obese groups experienced the largest moments when they carried the *Light* load and the smallest moments when they carried the *Heavy* load.

3.3.2. Unloaded side (stance phase)

In most cases, carrying a load resulted in an increased flexion moment ($p = 0.01$), an increased contralateral rotation moment ($p < 0.01$), and a decreased ipsilateral rotation moment ($p = 0.03$). However, both the obese and non-obese groups experienced increased ipsilateral (and decreased contralateral) obliquity moments while a load was carried. Considerably larger increases/decreases were observed among the non-obese groups ($p < 0.01$ and $p = 0.03$, respectively). Age resulted in a decrease in the

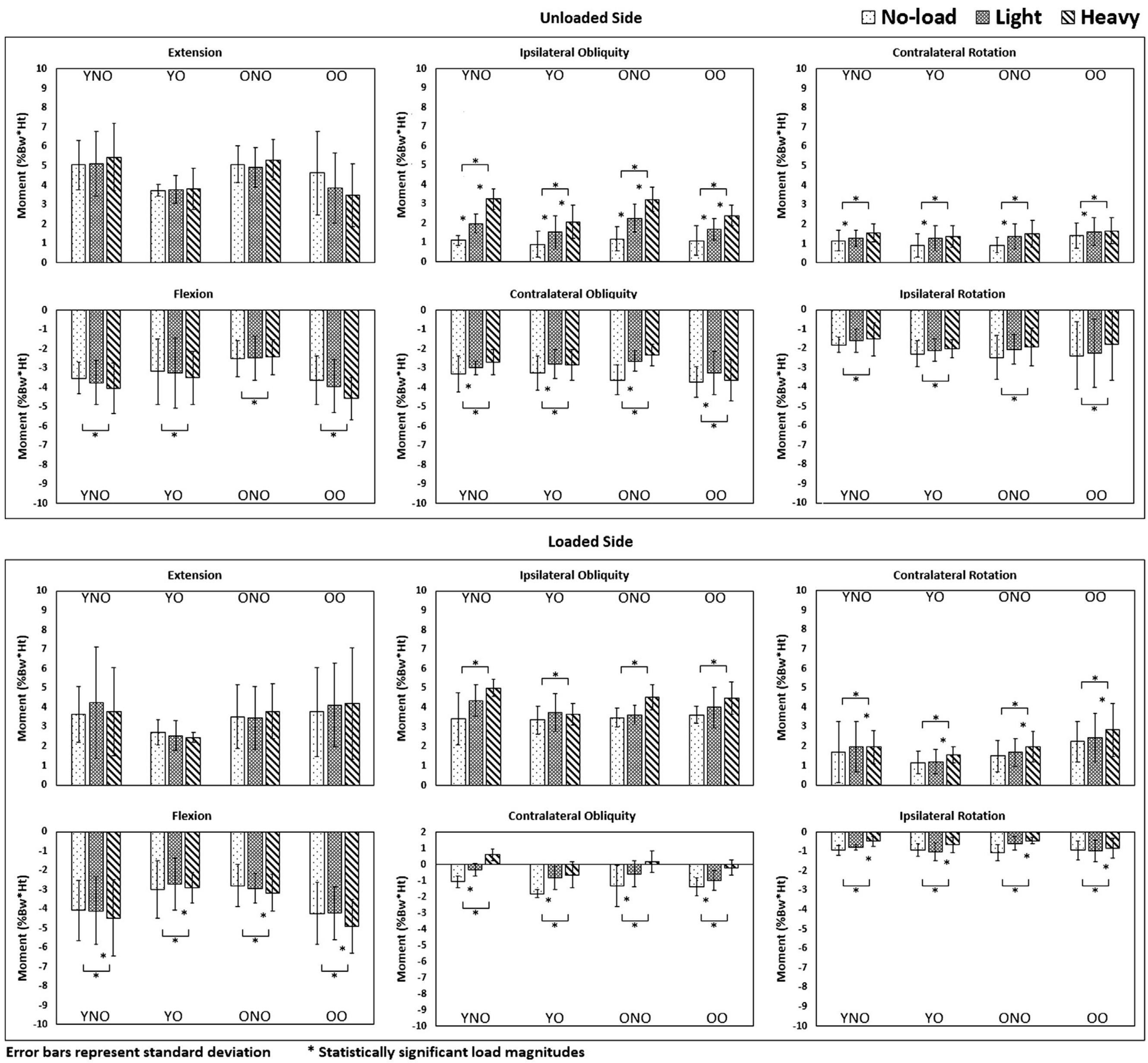

Fig. 3. Normalized peak L4/L5 moments at three load conditions. YNO: Young/Non-obese; YO: Young/Obese; ONO: Older/Non-obese; OO: Older/Obese.

extension moments among the non-obese groups, but an increase among the obese groups ($p = 0.01$). Age also resulted in a decrease in the ipsilateral rotation moment for the obese groups, but an increase among the non-obese groups ($p = 0.03$).

4. Discussion

Results of this study indicate that carrying a load in the dominant hand plays an important role in changing trunk kinematics and kinetics, but the effects are not dependent on age nor obesity category. Considering our hypotheses, the obese groups experienced greater absolute moments about the L4/L5 vertebral segment; however, these moments were mitigated when normalized to BW*Ht. Moments observed for the older groups were comparable to those observed for the young groups.

While load magnitude had a statistically significant effect on many of the tested absolute and normalized moments, increases in moments due to carrying the *Heavy* load across the four groups were smaller than the corresponding increases when participants walked with *No-load* (not exceeding 25.62 N m and 2.14%BW * Ht, respectively). However, the effects of such increases in moment on the internal loading of the spine remains unknown. In addition, the magnitude of the moments may be misleading while trying to understand the impact of dominant side one-handed carrying of a load. The ground reaction force is normally oriented towards (or closer to) the center of mass of the body, resulting in a small moment arm and, hence, a small moment. For instance, while a carried load in the dominant hand will cause two forces acting on the body side ipsilateral to the load, the weight of the upper body will cause another force acting on the contralateral side when people lean toward the contralateral side to maintain stability of their body. These forces have the potential to cause excessive loading of the lumbar spine, despite small moment magnitudes. Spinal loading is exacerbated with increased load magnitude since larger changes in trunk kinematics would be anticipated for maintaining balance of the body.

The obese groups experienced greater absolute moments than the non-obese groups during all the three load conditions (*No-load*, *Light*, and *Heavy*), although they carried a smaller percentage of their BW. Normalizing the moments to BW * Ht resulted in fewer changes in moments among the obese people due to load carrying compared to walking with *No-load*. In many cases, normalized moments among the obese groups were comparable to, if not smaller than, the corresponding moments among the non-obese groups. These findings suggest that obesity may not be a detrimental factor in affecting moments about the L4/L5 vertebral segment, particularly since neither obesity nor the interaction between obesity and any other independent variable had statistically significant effects on any of the 3D trunk angles. In fact, obese people could have a relative advantage during load carrying because of their heavier bodies (when solely considering the kinetics and kinematics of carrying). However, the impact of additional trunk weight may be applying an overall greater load on the spine relative to individuals with a healthy BMI.

Despite the statistically significant interaction of age and obesity on both the normalized extension and normalized ipsilateral rotation moments on the unloaded side, the maximum differences in moments between groups were practically negligible (1.43% BW * Ht and 0.51%BW * Ht, respectively). Consequently, carrying a load up to 10.21 kg in the dominant hand likely results in comparable trunk angles and moments about the L4/L5 vertebral segment for young and older working individuals up to 64 years of age. Therefore, older people may not be at an increased risk of MSDs during dominant side one-handed carrying of the *Light* and *Heavy* load due solely to kinetics and kinematics associated with the activity. Instead, other factors such as vertebral compression fracture due to aging (Old and Calvert, 2004) and muscle mass and strength decreases (Villareal et al., 2004; Beaufrere and Morio, 2000) may play a more important role.

Overall, dominant side one-handed carrying of a load of up to 10.21 kg had a substantial effect on the trunk motion of participants and may contribute to the development of MSDs during carrying tasks. This result supports the one-handed carrying maximum load recommendation of 10 kg for males provided in previous research (Lind and McNicol, 1968; Ganguli and Datta, 1977; Kilbom et al., 1992). Importantly, however, age and obesity do not appear to meaningfully exacerbate the effects of load on the kinematics and kinetics of the trunk. Obese people may actually have an advantage *in terms of kinematics and kinetics* while carrying a fixed load magnitude for short distances, since the load represents a smaller percentage of their BW than it does for non-obese individuals.

This research has two main limitations. First, the sample size in each participant group was relatively small. The experiment's duration and complexity impaired the ability to recruit more participants, particularly older individuals. Second, the tested loads were examined for short carrying distances (approximately 6 m). More research is needed to evaluate the effect of fatigue on the distribution of spinal loading during prolonged carrying tasks. Future research is also needed to examine the effects of one-handed carrying on ground reaction force among different age and obesity populations. It is assumed that ground reaction force will provide a better understanding of the change in the lumbar spine kinetics due to carrying a load in one hand. In addition, the simultaneous change in both trunk kinematics and kinetics during one-handed carrying needs to be studied. The present study focused mainly on the magnitude of change in kinetics and kinematics by examining peak trunk angles and moments. Although the present study provided a description of the change pattern in trunk angles along the gait cycle, the corresponding change in the moments about the L4/L5 vertebral segment remains understudied. Furthermore, while previous research has suggested that one-handed carrying increases hip muscle activity/joint forces on the side of the body contralateral to the carried load (Neumann and Cook, 1985; Neumann et al., 1992; Neumann, 1996; Bergmann et al., 1997), the effects of age and obesity were not the main objectives of these studies. In addition, to our knowledge, no work has been completed to examine these effects on the biomechanics of the knee or the ankle.

Declaration of Competing Interest

None.

Acknowledgments

The study was partially supported by the Deep South Center for Occupational Health and Safety, a National Institute for Occupational Safety and Health (NIOSH) Education and Research Center (Grant 2T42 OH008436-13). The contents are solely the responsibility of the authors and do not necessarily represent the official views of the Deep South Center or NIOSH. The authors would like to thank the individuals who participated in this study for their time.

Role of the funding source

The funding source had no involvement in the collection, analysis and interpretation of data; in the writing of the manuscript; or in the decision to submit the article for publication.

References

Andriacchi, T.P., Alexander, E.J., Toney, M.K., Dyrby, C., Sum, J., 1998. A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. J. Biomech. Eng. 120 (6), 743–749.
Andriacchi, T.P., Dyrby, C.O., 2005. Interactions between kinematics and loading during walking for the normal and ACL deficient knee. J. Biomech. 38 (2), 293–298.
Badawy, M., Schall Jr., M., Sesek, R., Gallagher, S., Davis, G., Capanoglu, M., 2018. One-handed carrying among older and obese individuals A systematic review to identify research gaps. Ergonomics 61 (10), 1345–1354. https://doi.org/10.1080/00140139.2018.1470680.
Beaufrère, B., Morio, B., 2000. Fat and protein redistribution with aging: metabolic considerations. Eur. J. Clin. Nutr. 54 (S3), S48–S53.
Bergland, A., Pettersen, A., Laake, K., 2000. Functional status among elderly Norwegian fallers living at home. Physiother. Res. Int. 5 (1), 33–45.
Bergmann, G., Graichen, F., Rohlmann, A., Linke, H., 1997. Hip joint forces during load carrying. Clin. Orthop. Relat. Res. 335, 190–201.
Blazek, K., Asay, J., Erhart-Hledik, J., Andriacchi, T., 2013. Adduction moment increases with age in healthy obese individuals. J. Orthop. Res. 31 (9), 1414–1422.
Bureau of Labor Statistics (BLS), 2015. Nonfatal occupational injuries and illnesses requiring days away from work, 2014. USDL-15-2205.
Chakraverty, R., Pynsent, P., Isaacs, K., 2007. Which spinal levels are identified by palpation of the iliac crests and the posterior superior iliac spines? Journal of Anatomy 210 (2), 232–236.
Cook, T., Neumann, D., 1987. The effects of load placement on the EMG activity of the low back muscles during load carrying by men and women. Ergonomics 30 (10), 1413–1423.
Corbeil, P., Plamondon, A., Handrigan, G., Vallée-Marcotte, J., Laurendeau, S. Ten, Have, J., Manzerolle, N., 2019. Biomechanical analysis of manual material handling movement in healthy weight and obese workers. Appl. Ergon. 74, 124–133.
Del Porto, H.C., Pechak, C.M., Smith, D.R., Reed-Jones, R.J., 2012. Biomechanical effects of obesity on balance. Int. J. Exerc. Sci. 5 (4), 301–320.
Dyrby, C.O., Andriacchi, T.P., 2004. Secondary motions of the knee during weight bearing and non-weight bearing activities. J. Orthop. Res. 22 (4), 794–800.
Finkelstein, E., Khavjou, O., Thompson, H., Trogdon, J., Pan, L., Sherry, B., Dietz, W., 2012. Obesity and severe obesity forecasts through 2030. Am. J. Prev. Med. 42 (6), 563–570.
Gallagher, S., Sesek, R., Schall, M., Huangfu, R., 2017. Development and validation of an easy-to-use risk assessment tool for cumulative low back loading: The Lifting Fatigue Failure Tool (LiFFT). Appl. Ergon. 63, 142–150.
Ganguli, S., Datta, S., 1977. Studies in load carrying in BK amputees with a PTB prosthesis system. J. Med. Eng. Technol. 1 (3), 151–154.

Grood, E., Suntay, W., 1983. A joint coordinate system for the clinical description of three-dimensional motions: Application to the knee. J. Biomech. Eng. 105 (2), 136–143.
Guelich, M., 1999. Prevention of falls in the elderly: A literature review. Top. Geriatric Rehab. 15 (1), 15–25.
Gu, J.K., Charles, L.E., Bang, K.M., Ma, C.C., Andrew, M.E., Violanti, J.M., Burchfiel, C. M., 2014. Prevalence of obesity by occupation among US workers: The National Health Interview Survey 2004–2011. J. Occup. Environ. Med. 56 (5), 516–528.
Hales, C.M., Carroll, M.D., Fryar, C.D., Ogden, C.L., 2017. Prevalence of obesity among adults and youth: United States, 2015–2016. US Department of Health and Human Services (USDHHS) No. 288, pp. 1–8.
Hoy, D., Bain, C., Williams, G., March, L., Brooks, P., Blyth, F., et al., 2012. A systematic review of the global prevalence of low back pain. Arthritis Rheum. 64 (6), 2028–2037.
Kilbom, A., Hagg, G., Kall, C., 1992. One-handed load carrying? Cardiovascular, muscular and subjective indices of endurance and fatigue. Eur. J. Appl. Physiol. 65 (1), 52–58.
Kouvonen, A., Kivimäki, M., Oksanen, T., Pentti, J. De, Vogli, R., Virtanen, M., Vahtera, J., 2013. Obesity and occupational injury: A prospective cohort study of 69,515 public sector employees. PLoS ONE 8, (10) e77178.
Lee, J., Nussbaum, M.A., 2013. Experienced workers may sacrifice peak torso kinematics/kinetics for enhanced balance/stability during repetitive lifting. J. Biomech. 46, 1211–1215.
Lerner, Z.F., Board, W.J., Browning, R.C., 2014. Effects of an obesity-specific marker set on estimated muscle and joint forces in walking. Med. Sci. Sports Exerc. 46 (6), 1261–1267.
Lind, A., McNicol, G., 1968. Cardiovascular responses to holding and carrying weights by hand and by shoulder harness. J. Appl. Physiol. 25 (3), 261–267.
Luckhaupt, S.E., Cohen, M.A., Li, J., Calvert, G.M., 2014. Prevalence of obesity among U.S. workers and associations with occupational factors. Am. J. Prev. Med. 46 (3), 237–248.
Marras, W.S., 2000. Occupational low back disorder causation and control. Ergonomics 43 (7), 880–902.
McGill, S., 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.
Nelson, A., Matz, M., Chen, F., Siddharthan, K., Lloyd, J., Fragala, G., 2006. Development and evaluation of a multifaceted ergonomics program to prevent injuries associated with patient handling tasks. Int. J. Nurs. Stud. 43 (6), 717–733.
Neumann, D., 1996. Hip abductor muscle activity in persons with a hip prosthesis while carrying loads in one hand. Phys. Ther. 76 (12), 1320–1330.
Neumann, D., Cook, T., 1985. Effect of load and carrying position on the electromyographic activity of the gluteus medius muscle during walking. Phys. Ther. 65 (3), 305–311.
Neumann, D., Cook, T., Sholty, R., Sobush, D., 1992. An electromyographic analysis of hip abductor muscle activity when subjects are carrying loads in one or both hands. Phys. Ther. 72 (3), 207–217.
Ng, M., Fleming, T., Robinson, M., Thomson, B., Graetz, N., Margonoet, C., et al., 2014. Global, regional, and national prevalence of overweight and obesity in children and adults during 1980–2013: a systematic analysis for the Global Burden of Disease Study 2013. Lancet 384 (9945), 766–781.
Old, J.L., Calvert, M., 2004. Vertebral compression fractures in the elderly. Am. Acad. Family Physic. 69 (1), 111–116.
Ortman, J., Velkoff, V., Hogan, H., 2014. An aging nation: The older population in the United States. United States Census Bureau, Economics and Statistics Administration, US Department of Commerce, 25-1140.
Peng, T., Pérez, A., Gabriel, K.P., 2018. The association among overweight, obesity, and low back pain in U.S. adults: A cross-sectional study of the 2015 National Health Interview Survey. J. Manipul. Physiol. Therap. 41 (4), 294–303.
Robertson, D.G.E., Caldwell, G.E., Hamill, J., Kamen, G., Whittlesey, S.N., 2013. Research methods in biomechanics. .
Roffey, D., Budiansky, A., Coyle, M., Wai, E., 2013. Obesity and low back pain: Is there a weight of evidence to support a positive relationship? Curr. Obes. Rep. 2 (3), 241–250.
Rohlmann, A., Zander, T., Graichen, F., Schmidt, H., Bergmann, G., 2014. How does the way a weight is carried affect spinal loads? Ergonomics 57 (2), 262–270.
Schulte, P.A., Wagner, G.R., Ostry, A., Blanciforti, L.A., Cutlip, R.G., Krajnaket, K.M., et al., 2007. Work, obesity, and occupational safety and health. Am. J. Public Health 97 (3), 428–436.
Seay, J., Selbie, W., Hamill, J., 2008. In vivo lumbo-sacral forces and moments during constant speed running at different stride lengths. J. Sports Sci. 26 (14), 1519–1529.
Sheng, B., Feng, C., Zhang, D., Spitler, H., Shi, L., 2017. Associations between obesity and spinal diseases: A Medical Expenditure Panel Study Analysis. Int. J. Environ. Res. Public Health 14 (2), 183.
Shojaei, I., Vazirian, M., Croft, E., Nussbaum, M.A., Bazrgari, B., 2016. Age related differences in mechanical demands imposed on the lower back by manual material handling tasks. J. Biomech. 49 (6), 896–903.
Tirosh, O., Sparrow, W.A., 2003. Identifying heel contact and toe-off using force plate thresholds with a range of digital-filter cutoff frequencies. J. Appl. Biomech. 19 (2), 178–184.
Toossi, M., 2015. Labor force projections to 2024: the labor force is growing, but slowly. Monthly Labor Rev., 1–32
United Nations Department of Economic and Social Affairs (UNDESA), 2015a. World population prospects: The 2015 revision, key findings and advance tables. Population Division, 1–59.
United Nations Department of Economic and Social Affairs (UNDESA), 2015b. World population ageing. Population Division, 1–149.
Villareal, D., Banks, M., Siener, C., Sinacore, D., Klein, S., 2004. Physical frailty and body composition in obese elderly men and women. Obes. Res. 12 (6), 913–920.
Wang, Y., Beydoun, M., Liang, L., Caballero, B., Kumanyika, S., 2008. Will all Americans become overweight or obese? Estimating the progression and cost of the US obesity epidemic. Obesity 16 (10), 2323–2330.
Wong, A., Karppinen, J., Samartzis, D., 2017. Low back pain in older adults: risk factors, management options and future directions. Scoliosis Spinal Disorders 12 (14), 1–23.
World Health Organization (WHO), 1995. Physical status: The use and interpretation of anthropometry: Report of a World Health Organization (WHO) Expert Committee. Switzerland, Geneva.
Wu, G., Siegler, S., Allard, P., Kirtley, C., Leardini, A., Rosenbaum, D., et al., 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. J. Biomech. 35 (4), 543–548.
Xu, Y.-W., Cheng, A.S.K., Li-Tsang, C.W.P., 2013. Prevalence and risk factors of work-related musculoskeletal disorders in the catering industry: A systematic review. Work 44, 107–116.
Zecevic, A., Salmoni, A., Speechley, M., Vandervoort, A., 2006. Defining a fall and reasons for falling: Comparisons among the views of seniors, health care providers, and the research literature. Gerontologist 46 (3), 367–376.