Ergonomics
Publication details, including instructions for authors and subscription information:
http://www.tandfonline.com/loi/terg20

Low back loads while walking and carrying: comparing the load carried in one hand or in both hands

Stuart M. McGill [a], Leigh Marshall [a] & Jordan Andersen [a]

[a] Spine Biomechanics Laboratories, Department of Kinesiology, University of Waterloo, 200 University Ave West, Waterloo, Ontario, Canada, N2L 3G1

Version of record first published: 05 Feb 2013.

Ergonomics, 2013
Vol. 56, No. 2, 293–302, http://dx.doi.org/10.1080/00140139.2012.752528

Low back loads while walking and carrying: comparing the load carried in one hand or in both hands

Stuart M. McGill*, Leigh Marshall and Jordan Andersen

Spine Biomechanics Laboratories, Department of Kinesiology, University of Waterloo, 200 University Ave West, Waterloo, Ontario, Canada N2L 3G1

(*Received 23 March 2012; final version received 9 October 2012*)

This study investigates the consequences of carrying load in one hand versus both hands. Six participants walked carrying buckets containing various weights. The weight was either carried in one hand or distributed evenly between both hands. Electromyography, force plate and body kinematic data were input to a three-dimensional anatomically detailed model of the spine to calculate spine loading. Carrying loads in one hand resulted in more load on the low back than when the load was split between both hands. When carrying 30 kg in one hand, the low back compression exceeded 2800 N; however, splitting the load between hands reduced low back compression to 1570 N (reduction of 44%). Doubling the total load by carrying 30 kg in each hand actually produced lower spine compression than when carrying 30 kg in one hand. Balancing the load between both hands when carrying material has merit and should be considered when designing work.

Practitioner Summary: Carrying a load in one hand (30 kg) resulted in more spine load than splitting the same load between both hands (15 kg). When carrying double the load in both hands (30 kg in each hand vs. 30 kg in one hand), spine load decreased, suggesting merit in balancing load when designing work.

Keywords: carrying; spine loads; electromyography; load carriage

1. Introduction

Walking and carrying are fundamental occupational activities as well as being essential to fulfilling activities of daily living. Although there are many techniques for carrying loads, the intent of this work was to investigate the consequences of carrying a load in one hand at the side of the body versus carrying load in both hands.

Most studies of carrying styles have focused on physiological energy costs. For example, Datta and Ramanthan (1971) reported that carrying loads in the hands was more physiologically demanding (in terms of heart rate, energy expenditure and oxygen consumption), while splitting the load between packs worn on the front and back of the torso was less demanding. Further, when carrying loads in the hands, carrying the load in one hand was more physiologically demanding than splitting the load between two hands. Interestingly, Legg (1985) noted that the limiting factor in carrying is not energy cost but rather the mechanical load tolerated by the musculature. Kilbom, Hagg, and Kall (1992) shared a similar impression from observations of fatigue in the forearm and hand muscles, together with cardiovascular fatigue during one-handed load carries.

Few mechanically based investigations of carrying loads in hands have been conducted. One of the few, by Bergmann et al. (1997), measured hip loads as a function of one- and two-handed carries, and noted higher loads with two-handed carries. In contrast, Neumann et al. (1992) recorded lower hip muscle activation with bilateral carries of loads at 20% of body weight compared to the load carried in one hand. Spine loading is influenced by different musculature and moment arm mechanics such that results from the hip should not be extrapolated to the spine. Observations of professional strongmen performing heavy carries in the ‘superyoke’ (the load is carried across the shoulders) and the ‘suitcase carry’ events (McGill, McDermott, and Fenwick 2008; the load is carried in one hand) highlighted the role of the guy-wire system formed by the torso musculature to support the spine and prevent buckling. This is a requirement for carrying these heavy loads together with the need to support the pelvis in the frontal plane to allow single leg stance and leg swing. In particular, the importance of the role of the abdominal wall and quadratus lumborum in assisting the hip musculature was illustrated. These muscles impose substantial loading to the spine. Traditional gait analysis often neglects the mechanics of the torso by considering it a rigid block rather than representing the spine as a bending beam under the control and support of a guy-wire system. Perhaps this is why quantification of spine loading has not been performed even though load distribution is a question often pondered in ergonomics.

*Corresponding author. Email: mcgill@healthy.uwaterloo.ca

This study assessed the spine loads in subjects carrying loads in one hand and in both hands to provide some guidance to decisions about carrying material. It was hypothesised that splitting the load between hands decreases spine load compared to carrying the load in one hand. It was further hypothesised that carrying a load in one hand creates more spine load than adding the same load to the other hand thereby doubling the weight carried.

2. Methods

Participants carried weighted buckets in their hands while electromyography (EMG) and external forces (force plates under the feet) were collected. Three-dimensional full-body kinematics were also tracked. These signals were input to an anatomically detailed model of the spine to assess spine loading.

2.1. Participants

Nine healthy male participants were recruited from the University of Waterloo to form a convenience sample. Initially, participants were screened and excluded from the study if they reported any previous or current low back pain/injury. This was a methodologically challenging data collection due to the three-dimensional (3D) full-body segment motion tracking, multi-channel EMG on both sides of the body and the need to land squarely on the force plate while carrying load. Further, the preparation of the EMG together with normalisation procedures required about 3 hours limiting the number of subjects and the actual walking trials that could be performed. Of the original nine participants, only six had full data from all channels to facilitate the involved data processing to calculate spine loading. These six participants had an average age of 22.7 ± 2.1 years, height 1.75 ± 0.10 m and weight 85.7 ± 16.7 kg. All participants read and signed a consent form prior to data collection that was reviewed by, and received ethics clearance through, the University Office of Research Ethics.

2.2. Instrumentation

2.2.1. Muscle activation from EMG

Twelve channels of EMG (Bortec Biomedical, Calgary, Canada, bandwidth 10–1000 Hz, common mode rejection ratio = 115 db at 60 Hz, input impedance = 10 GΩ) were collected by placing electrode pairs (Covidien Meditrace, Mansfield, MA, USA) over the following muscles on both sides of the body: right and left rectus abdominis (RRA and LRA) – 3 cm lateral to the navel; right and left external obliques (REO and LEO) – approximately 3 cm lateral to the linea semilunaris at the same level as the RRA and LRA electrodes; right and left internal oblique (RIO and LIO) – at the level of the anterior superior iliac spine and medial to the linea semilunaris, but superior to the inguinal ligament; right and left latissimus dorsi (RLD and LLD) – inferior to the scapula over the muscle belly when the arm was positioned in the shoulder mid-range; right and left upper (thoracic) erector spinae (RUES and LUES) – 5 cm lateral to the spinous process of T9, and right and left lumbar erector spinae (RLES and LLES) – 3 cm lateral to the spinous process of L3. Before the electrodes were adhered to the skin, the skin was shaved and cleansed with Nuprep™ (Weaver and Company, Aurora, CA, USA) abrasive skin prepping gel. Ag–AgCl surface electrode pairs were positioned with an inter-electrode distance of approximately 2.5 cm and were oriented in series parallel to the muscle fibres. The EMG signal was amplified and analogue to digital conversion was conducted with a 16-bit converter at a sample rate of 2160 Hz using the VICON Nexus™ (Los Angeles, CA, USA) motion capture system software.

Each participant performed a maximal contraction of each muscle for normalisation (Brown and McGill 2009). These techniques have been developed over 30 years in our lab to achieve isometric activation in ways that minimise the risk of back injury and muscle avulsion. Dynamic contractions create higher levels of motor unit activity according to known force–velocity relationships – these are incorporated into the modelling approach to estimate muscle force (McGill and Norman 1986). For the abdominal muscles (RRA, LRA, REO, LEO, RIO and LIO), each participant adopted a sit-up posture with the torso at approximately 45 degrees to the horizontal, with the knees and hips flexed at 90 degrees. Manually braced by a research assistant, the participant was instructed to produce a maximal isometric flexion moment followed sequentially by a right and left lateral bending moment and a right and left twisting moment. For the spine extensors (RLES, LLES, RUES and LUES) and latissimus dorsi, a resisted maximal extension in the Biering-Sorensen position was performed for normalisation. The latissimus dorsi muscles were cued by instructing the participants to pull their shoulder blades back and down during extension. The maximal amplitude observed during the normalising contraction for a specific muscle was taken as the maximal activation for that particular muscle.

2.2.2. Body segment kinematics and marker placement

Twenty-eight reflective markers for tracking linked segment kinematics were adhered to the skin with hypoallergenic tape over the following landmarks: right and left lateral malleoli, right and left medial malleoli, right and left calcanei, right and left medial femoral condyles, right and left lateral femoral condyles, right and left greater trochanters, right and left lateral iliac crests, right and left acromia, right and left medial epicondyles of the elbow, right and left lateral epicondyles of the elbow, right and left radius styloid processes, right and left ulnar styloid processes, right and left ear lobes, C7 vertebra and sternum. Fifteen rigid bodies moulded from splinting material were adhered to the skin with hypoallergenic tape over the following segments: right and left feet, right and left shins, right and left thighs, sacrum, T12, head, right and left upper arms, right and left forearms, and right and left hands. Four reflective markers were adhered with tape to each rigid body. The VICON Nexus™ motion capture system tracked the 3D coordinates of the reflective markers during the various trials at a sample rate of 60 Hz.

2.2.3. Force plates for external force measurement and kinetic analysis

Force plate (AMTI, Watertown, MA, USA) data were also collected using the VICON Nexus™ motion capture system and were sampled at the rate of 2160 Hz.

2.3. Exercise description

The participants were instructed to walk carrying buckets containing various weights (see Figure 1). The weight was either distributed evenly in two buckets held in either hand or in one bucket carried in the right hand. The only instruction the participants received was to try and land their left foot onto the force plate. There were seven walking conditions: 5 kg in both hands (5BH), 10 kg in one hand (10OH), 10 kg in both hands (10BH), 20 kg in one hand (20OH), 15 kg in both hands (15BH), 30 kg in one hand (30OH) and 30 kg in both hands (30BH). The order of conditions was randomised and each participant performed two consecutive trials for each condition. The second trial of each condition was chosen for analysis, unless data were contaminated or deemed unusable (marker or EMG dropout, participant did not step completely on the force plate, etc.). One of the participants was unable to complete the 30BH carrying condition.

Figure 1. Photo of subject carrying buckets either in one hand or with two hands while striding onto force plates. EMG is recorded for the torso musculature, together with 3D body segment kinematics.

2.4. Data analysis

2.4.1. EMG to capture muscle activation for the spine model

The EMG data were band-pass filtered between 20 and 500 Hz, full wave rectified, low-pass filtered with a second-order Butterworth filter at a cut-off frequency of 2.5 Hz (to mimic the frequency response of torso muscle discussed by Brereton and McGill [1998]), normalised to the maximal voluntary contraction of each muscle and down sampled to 60 Hz using custom LabVIEW™ (National Instruments, Austin, TX, USA) software.

2.4.2. Kinetic and kinematic model data to predict back loads

The 3D coordinates of the markers were entered into a software package (Visual3D™, C-Motion, Germantown, MD, USA), which calculated the spine angle as well as the reaction moments and forces about the L4/L5 joint. Normalised EMG signals and lumbar spine position data were entered into an anatomically detailed model of the lumbar spine. Specifically, the modelling process proceeded in four stages:

(1) The 3D coordinates of the joint markers drove a linked segment model of the arms, legs and torso constructed with Visual3D™ (C-Motion). This package output the lumbar spine postures described as three angles (flexion/extension, lateral bend and twist), bilateral hip angles and bilateral knee angles together with the reaction moments and forces about the L4/L5 joint.
(2) The reaction forces from the link segment model above were input into a 'Lumbar Spine model' that consists of an anatomically detailed, 3D ribcage, pelvis/sacrum and five intervening vertebrae (Cholewicki and McGill 1996). Over 100 laminae of muscle, together with passive tissues represented as a torsional lumped parameter stiffness element, were modelled about each axis. This model uses the measured 3D spine motion data and assigns the appropriate rotation to each of the lumbar vertebral segments (in reference with values obtained by White and Panjabi [1978]). Muscle lengths and velocities are determined from their motions and attachment points on the dynamic skeleton of which the motion is driven from the measured lumbar kinematics obtained from the subject. Also, the orientation of the vertebral segments along with stress/strain relationships of the passive tissues were used to calculate the restorative moment created by the spinal ligaments and discs. Some recent updates to the model include a much improved representation of some muscles documented by Grenier and McGill (2007).
(3) The third model, termed as the 'distribution-moment model' (Ma and Zahalak 1991; Guccione, Motabarzadeh, and Zahalak 1998), was used to calculate the muscle force and stiffness profiles for each of the muscles. This model uses the normalised EMG profile of each muscle along with the calculated values of muscle's physiological cross-sectional area, length and velocity of contraction to calculate the active muscle force and any passive contribution from the parallel elastic components.
(4) When input to the spine model, these initial muscle forces estimated from biological parameters are used to calculate a moment for each of the 18 degrees of freedom of the six lumbar intervertebral joints. However, the joint moments never exactly equal those predicted from the sum of muscle actions. The optimisation routine assigns an individual gain value to each muscle force to create a moment about the intervertebral joint that matches those calculated by the link segment model to achieve mathematical validity (Cholewicki and McGill 1994). The objective function for the optimisation routine is to match the moments with a minimal amount of change to the EMG-driven force profiles. The optimisation routine has a dynamic lower limit based on current activation, set on the optimised force output of the muscle to prevent any muscle from completely turning off. The adjusted muscle force and stiffness profiles are then used in the calculations of L4–L5 compression and shear forces.

In this way, the model was sensitive to the different muscle activation strategies and movement patterns of each subject. Both biological variability and mathematical/mechanical validity are accommodated. The average muscle activation, reaction and joint compression and shear forces (note that the L4/L5 joint was chosen as a representative joint to report these) as well as spine angles about the three axes of motion were also analysed from right toe off to just prior to right foot contact.

2.5. Statistical analysis

For data reduction purposes, the muscles were divided into four quadrants of the torso: left back (LLD, LUES, LLES), right back (RLD, RUES, RLES), left abdominal (LRA, LEO, LIO) and right abdominal (RRA, REO, RIO). Means and standard deviation of spine compression and shear and muscle activation were calculated for all conditions.

A priori contrasts were used to identify any significant differences in spine compression and shear between one-hand versus two-hand carrying methods when carrying the same total mass (i.e. 5BH vs. 10OH, 10BH vs. 20OH, 15BH vs.

Table 1. Spine compression (at L4/L5) for all walking trials while the left foot was in contact with the force plate and the right foot was completely off the ground.

Subjects	5 kg in both hands	10 kg in one hand	10 kg in both hands	20 kg in one hand	15 kg in both hands	30 kg in one hand	30 kg in both hands
S1	1082	1359	1326	2350	1713	3423	
S2	1108	1266	1237	1974	1386	2169	2150
S3	1114	1675	1465	1921	1732	2831	2387
S4	1786	1919	1805	2519	1933	3446	2523
S5	1237	1644	1360	2410	1618	3259	2484
S6	1024	1231	1234	2043	1382	2669	2149
Average	1225	1516	1404	2203	1628	2874	2339
SD	283.5	273.0	214.3	253.7	214.7	503.4	179.4

Note: Subject 1 was unable to perform the 30 kg carry in both hands. The comparison between 30 kg in one hand and 30 kg in two hands was conducted just between the five subjects who were able to do this.

30OH; $\alpha < 0.017$). The compression and shear data were tested for normality with the Shapiro–Wilk test and for homogeneity of variance with the Levene test using IBM® SPSS® Statistics 20 (IBM, New York, USA).

A repeated measures analysis of variance was conducted to test differences in load between hands and as more load was added to the hands (i.e. three loads and two carry conditions). As one subject was unable to carry 30 kg in each hand, the remaining successful subject scores for the 30 kg load in one and in both hands were compared using a paired *t*-test.

3. Results

3.1. *Hypothesis 1: splitting the load between hands decreases spine load compared to carrying the load in one hand*

All participants experienced greater spine compression when carrying the same total load asymmetrically compared to splitting the load evenly between both hands (Table 1). Further, a larger load magnified this effect ($p < 0.0001$). For example, carrying 10OH resulted in 23% more back load than carrying 5BH ($p < 0.001$), 20OH versus 10BH resulted in 57% more load ($p < 0.001$) and 30OH versus 15BH resulted in 82% more load ($p < 0.001$; see Figure 2).

Shear forces were found to be relatively low, given the upright walking posture, with all values below 520 N (see Table 2).

All muscle groupings of the torso followed the same trend as spine compression: there was greater activation during the asymmetric carrying conditions compared to symmetric carrying when the total load was even (the one exception was the 5BH vs. 10OH comparison where right back muscle activation was equal for both conditions; Table 3).

3.2. *Hypothesis 2: carrying a load in one hand results in more spine load than carrying the same load in both hands that doubles the total load carried*

When participants were asked to carry a load in one hand (i.e. 10OH or 30OH) and carry double that load in both hands, but splitting the load evenly (i.e. 10BH or 30BH), not only was there no significant difference found in spine load ($p > 0.063$) in the 10 kg condition, but in fact they also experienced greater spine compression (535 N) when carrying half the total load in one hand in the 30 kg condition ($p = 0.026$; see Table 1). It was interesting to observe an average of 111 N increase in compression from 10BH to 10OH. This difference was further amplified with greater load, as compression increased by 535 N from 30BH to 30OH.

Muscle activation was greater during the one-hand carrying trials for all muscles, with the exception of the right back for the 10OH versus 10BH comparison (see Table 3).

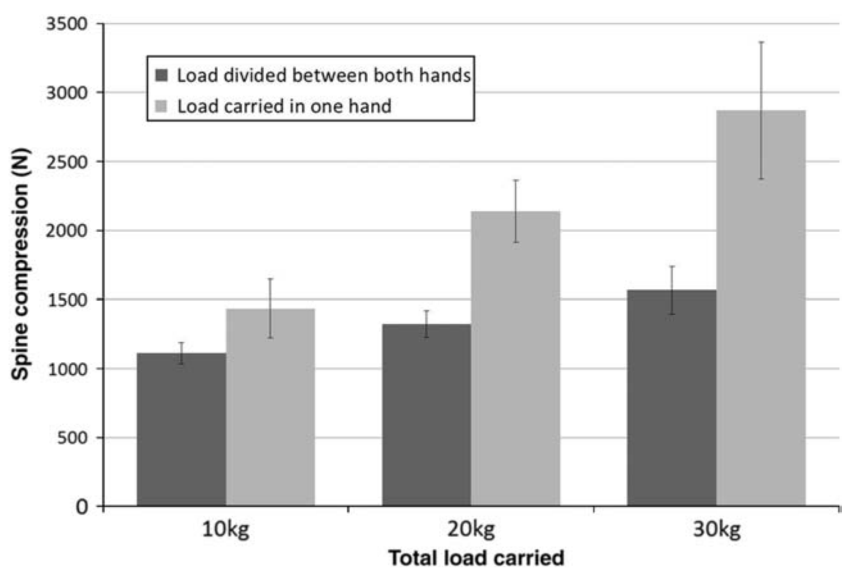

Figure 2. Compressive load on the lumbar spine as a function of load and whether it was carried in one hand or split between the two hands. The heavier the load, the larger the difference in spine loads between the hand conditions. Obviously, at low loads the torso mass dominates any differences between hand conditions but at higher loads the load dominates. Adding even more load would result in larger discrepancies.

Table 2. Spine shear (at L4/L5) for all walking trials while the left foot was in contact with the force plate and the right foot was completely off the ground.

Subjects	5 kg in both hands	10 kg in one hand	10 kg in both hands	20 kg in one hand	15 kg in both hands	30 kg in one hand	30 kg in both hands
S1	−19	35	−43	183	30	268	
S2	105	140	104	273	105	309	213
S3	173	234	173	276	235	424	313
S4	314	365	256	535	349	851	488
S5	234	200	170	370	239	652	348
S6	156	230	181	480	184	607	400
Average	167	200	154	353	190	519	294
SD	113.2	109.6	102.0	134.5	111.9	224.3	102.1

Note: Subject 1 did not perform the 30 kg carry in both hands. Negative values denote posterior shear of L4 on L5. Averages were taken from absolute values of the shear measurements to compare magnitudes of the shear force.

Table 3. EMG averages for all walking trials collected while the left foot was in contact with the force plate and the right foot was completely off the ground.

	5 kg in both hands		10 kg in one hand		10 kg in both hands		20 kg in one hand		15 kg in both hands		30 kg in one hand		30 kg in both hands	
	Average muscle activation (% MVC)	SD	Average muscle activation (% MVC)	SD	Average muscle activation (% MVC)	SD	Average muscle activation (% MVC)	SD	Average muscle activation (% MVC)	SD	Average muscle activation (% MVC)	SD	Average muscle activation (% MVC)	SD
LLD	1.8	2.4	8.8	10.3	2.8	2.4	19.9	15.3	4.5	4.0	28.9	19.4	6.7	5.8
LUES	2.4	2.5	6.8	5.2	3.0	1.4	12.7	8.2	4.8	2.4	20.2	11.8	7.1	2.8
LLES	3.9	3.1	7.9	2.4	4.3	2.3	13.7	5.2	5.1	2.2	25.3	10.4	8.4	3.8
RLD	2.8	3.8	2.9	3.8	3.3	3.8	6.0	2.9	5.7	6.0	7.3	4.9	4.9	3.4
RUES	2.2	1.3	2.1	0.6	3.0	0.9	3.8	1.6	4.5	1.0	6.2	3.7	7.3	3.1
RLES	4.2	2.3	4.4	1.9	4.7	2.3	7.3	3.9	6.9	3.0	11.7	6.2	10.3	5.2
LRA	1.4	1.4	4.3	4.9	2.5	1.8	14.5	12.0	2.5	1.5	26.0	24.1	3.9	3.8
LEO	2.6	2.5	10.2	7.8	3.2	1.9	27.1	9.6	3.6	2.2	47.3	29.9	6.1	4.1
LIO	6.9	4.7	10.3	8.0	7.9	4.7	23.5	17.3	9.5	6.5	32.5	24.5	18.5	11.5
RRA	0.8	0.6	3.6	4.2	2.0	1.3	10.3	13.3	1.8	0.7	20.8	27.7	2.8	1.7
REO	1.8	0.8	3.4	3.3	2.8	1.2	6.6	7.4	2.8	1.1	12.7	16.5	6.2	2.6
RIO	4.4	1.9	8.7	3.9	4.8	2.2	15.6	9.4	6.0	2.7	30.2	16.4	15.6	9.5
Left back	2.7	2.6	7.8	5.5	3.3	1.6	15.4	8.4	4.8	2.4	24.8	11.7	7.4	3.2
Right back	3.1	2.1	3.1	1.7	3.7	1.6	5.7	1.9	5.7	2.5	8.4	4.2	7.5	3.5
Left abdominal	3.7	2.8	8.3	6.7	4.5	2.5	21.7	10.7	5.2	3.3	35.3	24.1	9.5	6.1
Right abdominal	2.3	0.8	5.2	3.2	3.2	1.3	10.8	9.6	3.6	0.8	21.2	18.2	8.2	3.2

Note: MVC, maximal voluntary contraction. Right back is an average of RLD, RUES and RLES. Left back is an average of LLD, LUES and LLES. Right abdominal is an average of RRA, REO and RIO. Left abdominal is an average of LRA, LEO and LIO.

4. Discussion

Carrying loads in one hand results in substantially more compressive load on the low back than when the load is split evenly between the two hands. The resulting shear loads were relatively small and probably not of consequence, and the discussion is directed towards the compressive load. The first hypothesis, that splitting the same total load between hands reduces spine load, was confirmed. The second hypothesis, that doubling the total load by carrying the same load in both hands versus only one hand did not increase spinal load, was also proven. In fact, with heavier weights (the 30 kg condition), carrying double the load or 30 kg in each hand resulted in less spine compression than when carrying 30 kg in only one hand. It appears that the need to support a reaction moment from an asymmetric load is expensive in terms of torso muscle activity and the resulting spine load. In fact, this appears to dominate additional muscle activity needed to support the spine with a balanced but higher total load. Further, the enhanced loading due to one-handed carries appears to be magnified with larger weights. This is probably due to the increasing proportion of the load to the body mass ratio as the body weight remains fixed. Greater load will exacerbate the relationship, which is seen as a rising trend in Figure 2.

Although the question as to the cost of carrying a load in one hand versus two hands arises in ergonomics, no previous studies addressing back loading could be found for comparison to the data of this study. Cook and Neumann (1987) did note that back muscle activity increased when the load was carried in one hand versus two hands. Perhaps the most relevant data for back load were those observed in professional strongmen who perform the ‘suitcase carry’ as an event, together with the ‘yoke carry’, where the load is projected down the spine. Analysis of the ‘yoke carry’ by McGill, McDermott, and Fenwick (2008) suggested that lateral spine muscles, such as the quadratus lumborum and the lateral abdominal wall in particular, together with the erector spinae play an important role in holding and stiffening the pelvis level to prevent the pelvis from bending towards the side of leg swing. Thus, these muscles support lateral spine moments together with the moment of the hip in abduction on the stance leg. This extra muscle activity is imposed on the spine accounting for the elevated load.

Interpretation of the data presented here is limited by the small sample size. While more subjects participated, the full 12-channel EMG electrode integrity was challenging to preserve over the full data collection of the normalisation and the carrying tasks. Nonetheless, there were six successful subjects with no missing data. Even so, the results of carrying a load in one hand were so dramatic that statistical significance supported the clinical significance.

In summary, carrying a load in one hand imposes substantially more spine load than if the load were to be split between two hands. This effect is magnified as the magnitude of the load carried increases. Even doubling the total load by carrying the weight in both hands did not increase the spine load, at heavier weights the spine load was lower. The recommendation to balance the load between both hands when carrying material appears to have merit and should be considered when designing work.

Acknowledgements

The authors gratefully acknowledge the financial support of the Natural Science and Engineering Research Council of Canada (NSERC).

References

Bergmann, G., F. Graichen, A. Rohlmann, and H. Linke. 1997. “Hip Joint Forces During Load Carrying.” *Clinical Orthopaedics and Related Research* 335: 190–201.

Brereton, L. C., and S. M. McGill. 1998. “Frequency Response of Spine Extensors During Rapid Isometric Contractions: Effects of Muscle Length and Tension.” *Journal of Electromyography and Kinesiology* 8: 227–232.

Brown, S., and S. McGill. 2009. “The Intrinsic Stiffness of the In Vivo Lumbar Spine in Response to a Variety of Quick Releases: Implications for Reflexive Requirements.” *Journal of Electromyography and Kinesiology* 19 (5): 727–736.

Cholewicki, J., and S. M. McGill. 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., and S. M. McGill. 1996. “Mechanical Stability of the In Vivo Lumbar Spine: Implications for Injury and Chronic Low Back Pain.” *Clinical Biomechanics* 11 (1): 1–15.

Cook, T., and D. Neumann. 1987. “The Effects of Load Placement of the EMG Activity of the Low Back Muscles During Load Carrying by Men and Women.” *Ergonomics* 30 (10): 1413–1423.

Datta, S., and N. Ramanthan. 1971. “Ergonomic Comparison of Seven Modes of Carrying Loads on the Horizontal Plane.” *Ergonomics* 14 (2): 269–278.

Grenier, S. G., and S. M. McGill. 2007. “Quantification of Lumbar Stability Using Two Different Abdominal Activation Strategies.” *Archives of Physical Medicine and Rehabilitation* 88 (1): 54–62.

Guccione, J., I. Motabarzadeh, and G. Zahalak. 1998. “A Distribution-Moment Model of Deactivation in Cardiac Muscle.” *Journal of Biomechanics* 31: 1069–1073.

Kilbom, A., G. Hagg, and C. Kall. 1992. “One-Handed Load Carrying – Cardiovascular, Muscular and Subjective Indices of Endurance and Fatigue.” *European Journal of Applied Physiology* 65: 52–58.

Legg, S. 1985. “Comparison of Different Methods of Load Carriage.” *Ergonomics* 28 (1): 197–212.

Ma, S., and G. I. Zahalak. 1991. “A Distribution-Moment Model of Energetics in Skeletal Muscle.” *Journal of Biomechanics* 24 (1): 22–35.
McGill, S. M., and R. W. Norman. 1986. “The Volvo Award for 1986: Partitioning of the L4/L5 Dynamic Moment Into Disc, Ligamentous and Muscular Components During Lifting.” *Spine* 11 (7): 666–678.
McGill, S., A. McDermott, and C. Fenwick. 2008. “Comparison of Different Strongman Events: Trunk Muscle Activation and Lumbar Spine Motion, Load and Stiffness.” *Journal of Strength and Conditioning Research* 23 (4): 1148–1161.
Neumann, D. A., T. M. Cook, R. L. Sholty, and D. C. Sobush. 1992. “An Electromyographic Analysis of Hip Abductor Muscle Activity When Subjects are Carrying Loads in One or Both Hands.” *Physical Therapy* 72 (8): 207–217.
White, A., and M. Panjabi. 1978. “The Basic Kinematics of the Human Spine. A Review of Past and Current Knowledge.” *Spine* 3: 12–20.