2014 5th IEEE RAS & EMBS International Conference on
Biomedical Robotics and Biomechatronics (BioRob)
August 12-15, 2014. São Paulo, Brazil

Second Spine: a Device to relieve stresses on the upper body during loaded walking

Joon-Hyuk Park, Damiano Zanotto, Vineet Vashista, Xin Jin, Paul Stegall and Sunil K. Agrawal*

***Abstract*— The target of this work is to experimentally validate the *Second Spine*, a wearable device recently developed by our group to transfer forces from shoulder to pelvis during loaded walking. A key-feature of the Second Spine compared to traditional framed backpacks is the adjustable stiffness of its structure, which allows the wearer to change the load-bearing behavior of the device. In line with previous studies on loaded walking, we investigate biomechanical and physiological variables on a small group of young healthy subjects, as they walked on a treadmill under 3 different conditions: free walking, walking with a backpack of 25% of subject's Body Weight (BW), and walking with the same backpack while wearing the device. Results indicate that wearing the Second Spine significantly reduces the pressure on shoulders and induces smaller deviations from unloaded walking in terms of gait timing and stride length. The activations of the rectus femoris and the gastrocnemius muscles, along with the kinematics of the knee joint, provide indirect evidence that dynamic loads were rigidly transmitted from the shoulder to the waist. We discuss how these preliminary findings might be relevant for the prevention of injuries related to load carriage, and how they set important guidelines for the next generation of the Second Spine.**

I. Introduction

Many works have been published in the past, which deal with biomechanical and physiological effects of human load carriage. The rationale behind studies focusing on this topic relies not only on the design guidelines for novel, more efficient load carrying systems, but also from the necessity to reduce the risk of injuries associated with prolonged load carriages. Among these injuries, the most common ones involve the lower limbs (foot blisters, metatarsalgia, stress fractures, knee pain) and the lower back (low back pain, spasm, disc tear/herniation, spinal stenosis) [1]. Prevalence of low back pain is higher for subjects that carry heavy loads on a regular basis (e.g. material handlers and nurses [2], infantrymen [3] and young scholars [4]). Adolescents, for example, show significant effects of muscle fatigue after just 15 minutes of walking with rucksack loads of 15% - 20% BW [5], while exceeding the latter range has been associated with back pain [6].

Besides load magnitude, load location plays a role in determining posture and efficiency of load carrying, ultimately affecting occurrence of injuries and pain [7]. It has been shown that locating the load mass close to the body Center Of Mass (COM), e.g., by means of double-packs, results in lower metabolic cost [8] and lower postural deviations from natural walking [9]. Less forward lean of the trunk reduces the incidence of back problems [10]. However, double-packs hinder arm and trunk movements more than back-packs, they may restrict the field of vision and even induce ventilatory impairments and heat stress symptoms [11], [12]. Therefore, back-packs are currently the most common and versatile solutions for scholars, recreational hikers and military infantrymen.

Subjects walking with a backpack increase forward lean of the trunk to maintain the combined COM of upper body and carried load over the feet. This increased forward lean has been hypothesized to cause foot strain and injury [1] and back injuries [13]. By locating loads higher in the pack, this postural adaptation is reduced because less rotation of the trunk is required to bring the COM of the backpack over the feet [14]. However, since Head, Arms and Trunk (H.A.T.) can be roughly modeled as an inverted pendulum rotating about the hip joint, such load arrangements may actually result in higher upper body inertia and consequently higher muscle activations at the pelvis to keep postural stability, especially while walking on uneven terrain [14].

Carrying a backpack also affects the gait kinematics. Previous works have reported higher stance-phase peak knee flexion, reduced swing phase, longer double support phase, and increased ankle dorsi/plantar flexion when subjects carried a load [1], [7]. Overall, these strategies help the body damp from increased impact forces due to the added mass at initial contact and transfer the load between the legs during weight acceptance.

Backpacks featuring a frame and a hip belt have been shown to alleviate stress to the shoulders by partially transferring the load to the hip [15], even though their effectiveness depends on the specific backpack model. Decreased pressures on the shoulder may not only reduce shoulder discomfort and nerve compression causing rucksack palsy [15], but also reduce the stress to the spine, since the rigid structure of the pack acts as an alternative pathway to transfer loads to the lower body. However, there are situations where significant loads have to be carried outside the backpack (e.g., a military tactical vest weighs up to 13.6kg, nearly 18% BW [16]); in these cases, a more *wearable* solution to transfer loads from shoulders to pelvis is desirable, and indeed this was the main motivation behind the design of the Second Spine. To overcome some of the drawbacks of double-packs and backpacks, new load-carrying solutions have been proposed to transfer loads from shoulders and back to the waist without the need of a backpack frame [17], [18].

This paper deals with the *Second Spine*, a passive, wearable load carrying device which was designed to transfer loads from shoulders to pelvis. Its unique feature is the

J. H. Park, D. Zanotto, V. Vashista, X. Jin, P. Stegall and S. K. Agrawal (Sunil.Agrawal@columbia.edu) are with the Dept. of Mechanical Engineering, Columbia University, NY 10027, USA.

Fig. 1. Cone-shaped joints allow two segments to share the maximum area of contact when they are pulled together by a cable routed from the top segment to the bottom. Springs are interposed between two mating segments to ensure separation when the cable is not tensioned (left). Prototype of the Second Spine (right) [18].

capability to change its load-bearing characteristics from high stiffness to high flexibility/compliance by means of a simple manual adjustment that can be operated by the user while he/she is wearing the device. We first introduce the design of the system, then present experimental results aimed to assess the effects of the device on the user during loaded walking, compared to a traditional framed backpack. In the last part of the paper, we discuss the potential impact of our preliminary findings on the prevention of low back injuries and other load-related injuries.

II. Methods

A. Design of the device

The Second Spine was designed to provide an alternate load pathway to the human spine in an effort to reduce stress on the upper body and therefore mitigate back and shoulder injuries. The secondary load pathway forms in parallel with the spine, so that some of the load may still be borne by the wearer. The device consists of three columns, two anterior and one posterior (Fig. 1, right side). The columns act as compliant members in parallel with the spine. The amount of load the wearer's spine has to support depends on the combined stiffness of the columns and the gap between the wearer's shoulder and the device in the unloaded configuration [18].

This device was intended to be worn close to the body and not to be taken off when the load is removed. Therefore, flexibility had to be guaranteed in the no load configuration, so that the wearer's movement was not constrained. This was accomplished through the use of cone shaped segments (Fig. 1, left side). These segments have an external and internal cone that are constrained by springs to maintain separation of adjoining segments. This allows for constrained rotations of the columns about the horizontal axes, which in turn allow trunk tilt, obliquity, and rotation. When the device

Fig. 2. An illustration of the experiment protocol. The order of sessions was randomized among subjects

is to be loaded, cables that route through each column can be tensioned to compress the springs, bringing the internal and external cones of consecutive segments in contact, thereby locking the joint. This design allows the device to be either rigid or flexible through the adjustment of the cables using a turn buckle mechanism mounted at the bottom of each column. Size adjustments are available to fit different torso lengths, waist circumferences, chest breadths, and shoulder slopes, so as to cover 25th to 75th percentile of male population. The advantage of this device over traditional framed backpacks is that it can support loads that are not usually carried in backpacks, such as protective or equipment vests, in addition to backpack loads. The total system weighs 1.8 kg including foam-paddings added for user comfort.

B. Experimental protocol

Six healthy male adults participated in this pilot study (age 28±3 years, height 1.80±0.03 m, weight 78.8±10.3 kg). All subjects were free of any physical disorder that may impede their walking capability. Subjects completed the full protocol during a single session. The study (Fig. 2) consisted of three 15-minute sessions: (a) baseline session (BL): subjects walked on a treadmill while wearing a 1.5 kg tactical vest, without backpack or Second Spine; (b) session 1 (S1): subjects walked with the tactical vest, and a loaded backpack (25 % BW) but without the Second Spine; (c) session 2 (S2): subjects walked with the same backpack and a vest as well as the Second Spine, which was fitted under the vest. Each participant completed the BL first, then S1 and S2 in randomized order. The speed of the treadmill was set to 1.25 m/s (2.24 MPH) for all sessions, and 25-min breaks were given between sessions.

Resistive pressure sensors (Tekscan® F-Scan VersaTek Wireless, Sensor-3000E) were placed on both shoulders to measure the transferred loads. It is comprised of 954 sensing elements (sensels) with resolution of 3.9 sensels per cm^2. Each sensor was sandwiched with 0.2 in (5 mm) polyethylene foams (Plastazote®) for better load distribution and to protect the sensor from crinkling. To compensate for drift during the measurement, a three-point power calibration was performed before and after each session [19].

Markers were placed on the human body to measure gait kinematics. A VICON® motion capture system with ten cameras (Model: T40S) was used to track these markers. The marker locations were adapted from Kadabah et. al.'s work [20]. Upper body markers were placed on the vest, and pelvis markers were placed on the hip brace.

Surface Electromyography (SEMG) was used to measure activations of 5 muscles: gastrocnemius medialis (GM), rectus femoris (RF), biceps femoris (BF), trapezius (TRAP), and erector spinae thoracis (ES). All muscles were recorded

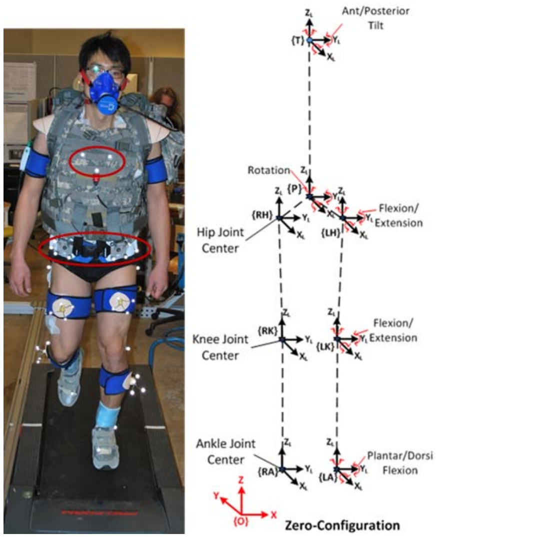

Fig. 3. A subject wearing the Second Spine, tactical vest and backpack during a test (left). Global and joint coordinate systems used in kinematic analysis (right).

from the right side of the body. Before attaching surface electrodes, electrode sites were shaved and cleaned with rubbing alcohol. Wet gel, Ag/AgCl disposable surface electrodes (Blue sensor N-00-S, Ambu A/S, Ballerup, Denmark) with inter-electrode distance of 20 mm were placed on the muscles according to the SENIAM guidelines [21] and were left in place throughout the duration of the test. Single-differential signals were high-pass filtered with a 1st order analog filter ($f_c = 10$ Hz), digitalized and sent to a wireless desktop unit (DTS Desktop Receiver, Noraxon Inc., Scottsdale, Arizona) which was connected to the VICON digital acquisition board.

A MetaMax 3B (Cortex, Germany), cardiopulmonary testing system was used to measure the oxygen consumption rate ($\dot{V}O2$) and the heart rate. These were regarded as indicators of the metabolic cost of walking under the three different configurations. The airflow volume sensor was calibrated with a 3-liter syringe before testing of each subject. The oxygen sensor was calibrated with ambient air before each recording.

Kinematic and force data were sampled at a frequency of 100 Hz, while EMG data were sampled at 1.5 kHz. Force data were synchronized to EMG and kinematic data by means of an external trigger/synch signal.

C. Data Analysis

Force on shoulders, kinematic data and muscle activations recorded during the last 60 seconds of each session were included in the analysis. Forces measured by the resistive sensors on the left and right shoulders were averaged over the one-minute-long time window, summed and normalized to the backpack weight.

Marker data were analyzed using VICON Nexus and MATLAB (The MathWorks Inc., Natick, Massachusetts) to estimate the joint angles. Global and local coordinate systems were those illustrated in Fig 3, right side. The vest and the hip brace fitted snugly on the body, thereby minimizing relative motions during walking. Trunk and hip angles were

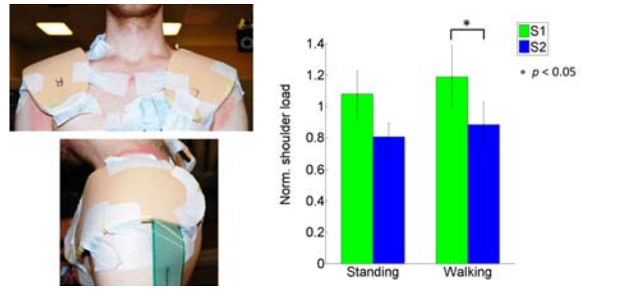

Fig. 4. Pressure sensors placed on shoulders of a subject (left). Average Total Force acting on the shoulders, normalized to backpack weight (right). Error bars indicate $\pm$ SE.

calculated with respect to the inertial frame instead of the pelvis frame, as pelvis kinematics were not directly obtained from the pelvis anatomical landmarks. Knee angles were calculated from the relative locations of shank and thigh frames. Similarly, ankle angles were calculated from the relative locations of foot and shank frames. Gait events (toe off and heel strike) were calculated for each leg using anterior-posterior displacements of toe and heel markers relative to the sacrum marker, as suggested in [22]. Based on these events, stride length and timing variables (i.e., swing period, stance period, and cadence) were computed within each stride and then averaged. Stride length was normalized to subject's height prior to statistical analysis.

Upper body raw EMG signals were processed for ECG noise reduction through adaptive filtering (Noraxon Myoresearch XP). An auxiliary EMG channel was recorded over the left Pectoralis Major to get a copy of the noise and check the effectiveness of the filter. Then, all signals were post-processed using custom MATLAB code: after band-pass filtering (2nd order Butterworth, 20-500 Hz) and full-wave rectification, signals were split into gait cycles based on the gait events and integrated over single gait cycles (iEMG). Prior to statistical analysis, iEMG data of each subject were normalized to the peak values recorded during the baseline session.

Both $\dot{V}O2$ and a heart rate were averaged over the duration of each session, discarding the first and the last minute of each session. The average oxygen consumption rate was normalized on subject's weight prior to statistical analysis.

Statistical analysis was conducted on each metric using SPSS (IBM Corp, NY, USA). Pairwise comparisons were performed with Wilcoxon signed-rank tests [23] to check for statistically significant effects ($\alpha < 0.05$) of the three walking modes BL, S1 and S2 on the biomechanical and physiological variables described above.

III. Results

Figure 4 (right) illustrates the average normalized force acting on the shoulders. Prior to S1 and S2, subjects were asked to stand still for 30 seconds, while pressure sensors measured the static force acting on their shoulders. Those data are labeled as "Standing" in Fig. 4. Wearing the Second Spine reduced the force transmitted to the shoulders, the effect being significant in the dynamic case ($p = .043$) but only close to significance in the static case ($p = .068$).

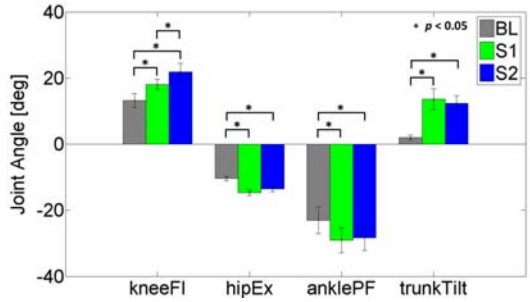

Fig. 5. Average sagittal plane joint angles for different sessions. Error bars indicate ± SE.

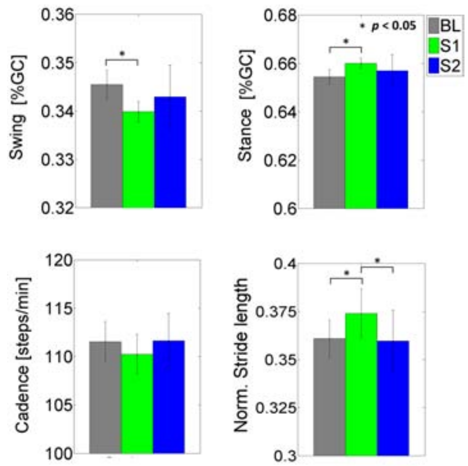

Fig. 6. Average of gait timing parameters and stride length. Error bars indicate ± SE.

Figure 5 presents the sagittal plane joint angles for the three sessions, averaged over the subjects. Each plot shows the average peak angle measured during the full gait cycle, except for the knee flexion, which captures the average peak measured during weight acceptance. Compared to BL, subjects showed significant increases in the knee flexion angle, both in S1 and in S2. Interestingly, a significant increase in the knee flexion angle was observed between S1 and S2. Similarly, the peak hip extension angles, the peak ankle plantar flexion angles and the peak trunk forward tilt angle were significantly higher in sessions S1 and S2 compared to BL.

Gait timing variables and stride length are shown in Fig 6. Carrying a backpack load (S1) significantly increased the stance period compared to free walking (BL) and, in turn, decreased the swing period. In a similar way, the average stride length significantly increased from BL to S1. Interestingly, all these deviations were attenuated (i.e., parameters changed back, approximately, to the their baseline values) when subjects carried the same backpack load, but wearing the Second Spine (S2). This effect was more marked for the stride length, with a significant decrease detected between the average value measured in S1 and the one measured in S2. Compared to the stride length, cadence showed an opposite trend. This result was expected since the product of the two is proportional to the walking speed, and the latter was constant throughout all the sessions. However, changes in cadence did not reach significance.

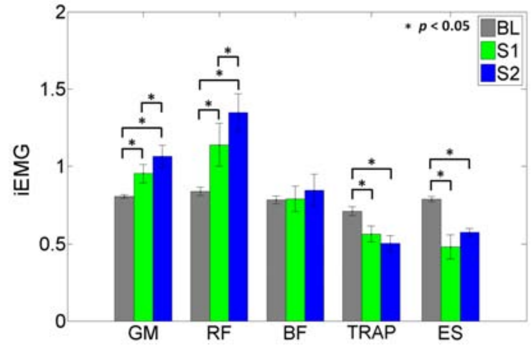

Fig. 7. Average integrated EMG measured on 5 muscles. Prior to compute group averages, signals were normalized on the peak baseline values. Error bars indicate ± SE.

Figure 7 shows the average iEMG values of the 5 recorded muscles across the walking conditions BL, S1 and S2. GM and RF showed higher muscle activations when subjects walked with a backpack load, reflecting increased efforts to provide a power-burst at push-off (GM) and to control knee flexion during weight acceptance (RF). Indeed, pairwise comparisons showed that the average BL activation was significantly lower than both the S1 and the S2 activations in both muscles. In addition, activity of both GM and RF were significantly higher during S2 than they were during S1. The activity of the BF was little affected by walking mode ($p > .1$). The activity of the TRAP muscle, which is responsible for stabilization and upward rotation of the scapula, showed a significant reduction in both S1 and S2 compared to BL. The ES muscles control extension of the vertebral column with respect to pelvis. The muscle activation of ES was significantly reduced in both S1 and S2, compared to BL.

The average $\dot{V}O2$ and a heart rate were significantly higher during S1 and S2 than they were during free walking ($p < .01$), thus reflecting subjects' increased metabolic cost when they carried the backpack load. No significant differences were observed between S1 and S2.

IV. Discussion

Data recorded by pressure sensors indicated that wearing the Second Spine in loaded walking significantly reduced the pressure on the shoulders, thereby possibly relieving stress on spine and torso from backpack load. Large stress in the shoulders has been related to various forms of soreness and injuries, such as back pain, rucksack palsy, and pressure ulcers [1], [7], [24]. Therefore, this result is promising and might indicate the potential of our device to reduce discomfort and musculoskeletal injuries during loaded walking.

Relieving stress from the shoulders was not the only effect of wearing the Second Spine. When walking with the backpack (S1), subjects significantly decreased the swing period, correspondingly increased the stance period, and increased the stride length. These adaptations, however, were not detected when subjects wore the Second Spine (S2).

Previous findings on loaded walking regarded such adaptations as a strategy of the human motor system to increase postural stability [25]–[27]. Indeed, carrying a backpack load shifts the COM of body-plus-pack posteriorly. This alteration in the mass distribution requires additional time to shift the COM over the base of support [28]. Thus, longer stance period helps shifting the load from one leg to the contralateral leg during mid-stance, thereby enhancing body stability during load carriage. The increased stride length measured in S1, which was favored by a significantly larger hip flexion angle, also contributed to a longer stance period.

Conversely, the fact that swing period, stance period, and stride length were close (i.e., not significantly different) to the unloaded walking values when subjects wore the device might indicate improved body stability caused by the device. We hypothesize that this was due to the COM of body-plus-pack being shifted slightly higher ($\approx$ 3 cm) on the human body when subjects wore the device. There are at least two clues that support this hypothesis: the slight decrease in trunk tilt and the slight increase in the activation of the ES measured in S2, compared to their corresponding values in S1. Subjects walking with a backpack increase forward lean of the trunk to maintain the combined COM of upper body plus carried load over the feet. Decreased trunk tilt, then, suggest smaller postural changes to keep the COM over the base of support. This, in turn, may happen if the COM is shifted upward, since less rotation is required to bring the COM of the backpack over the feet [14]. Despite this postural strategy, it has been observed that the overall COM is still located rearward compared to natural walking [29], [30]. As a consequence, the trunk is subjected to an external extension moment which causes a reduction on the activity of the ES (spine extensor) similar to the one observed in session S1 [4]. Following a similar reasoning, then, one may deduce that the slight increase in the activation of ES measured in S2 compared to S1 was related to the reduced trunk tilt observed in the same session, the latter being induced by a higher COM. Overall, even though further experiments must be conducted in the future to better support this hypothesis, we found evidence that indicate the Second Spine may favor postural stability in loaded walking.

In line with previous studies on backpack load carrying, in S1 and S2 we observed significant increase in the stance-phase peak knee flexion angle [1], and corresponding significant increase in the hip extension of the contralateral leg [4] with respect to unloaded walking. These changes can be explained as a shock absorption strategy used by the human motor system during the double support phase. They indicate that external loads pose challenges to the motor system, especially during the transition from double support to single support phase. The increased activations of the RF found in S1 and S2 were necessary to control knee flexion (i.e., prevent the knee from buckling) at weight acceptance; higher activations of the GM provided larger power-burst to support propulsion of the body-plus-pack. These observations confirm the results of previous studies on loaded walking [1], [27], [30].

More interestingly, our data showed further increases in knee flexion angle and in the activations of GM and RF muscles when subjects wore the device (S2) compared to simple loaded walking (S1). During walking, the vertical acceleration of the H.A.T. follows an approximately sinusoidal pattern, whose frequency is twice that of the gait cycle [31]. Assuming that the backpack is rigidly attached to the upper body, this oscillatory motion induces extra dynamic forces that adds up to the static ones, and must be transferred to the ground through the wearer's body. Since the backpack load carried by the subjects was the same in S1 and S2, we hypothesize that these increasing trends in the activations of RF and GM were related to the Second Spine providing a secondary, stiff load path which acts in parallel to the vertebral column for transferring loads to the pelvis. Though relieving the spine from some load stresses, this may have the opposite effect on the lower extremities, due to peak dynamic forces that are no longer attenuated by the spine and the surrounding soft tissues. Higher dynamic loads acting on the body have been previously related to increased activations of triceps surae and quadriceps muscles [32]. There is evidence that the human bones and soft tissues act as shock absorbers [33], attenuating the magnitude of the periodic dynamic loading resulting from locomotion. The healthy spine has a resonant frequency of about 5Hz, while it can successfully attenuate frequencies above 15Hz [34]. This damping feature has been attributed to the intervertebral discs, acting as flexible links and allowing the spinal column, its musculature and ligaments to dissipate energy by bending [34]. Based on these observations, we hypothesize that the changes in the activations of GM and RF and in the knee flexion angle measured between S1 and S2 were due to an increase in the dynamic loads transmitted to the legs caused by the device. If this was the case, then providing the Second Spine with tunable stiffness/damping between the upper structure and the hip belt may enhance the effectiveness of our design.

Unexpected results were obtained in the activations of the TRAP muscle, which resists shoulder depression under the weight of the backpack. Interestingly, instead of increasing, the activation of this muscle decreased in S1 compared to unloaded walking. Also, the Second Spine reduced the activity of this muscle only slightly compared to S1, while it caused a significant reduction in the forces applied to the shoulders. Results on the effect of backpack loads on the activation of the trapezius are mixed, with some studies reporting a significant increase with respect to unloaded walking [15], [30], [35] and others reporting non-significant changes [4], [5], [27], [36] or slight decrease [29]. Activations of this muscle might have been biased by larger arm swing during BL (when arm motion was not hindered by the backpack) or by different positions of the arms adopted by the subjects across the sessions [29].

Lastly, all the other biomechanical and physiological variables did not show any significant difference between sessions S1 and S2, thus indicating that these parameters were not adversely affected by the Second Spine.

V. Conclusion

Prolonged pressure on shoulders may cause numbness, weakness and temporary paralysis. In this paper, we showed that wearing the Second Spine significantly reduced the pressure transmitted on the shoulders that is a risk factor for low back pain and foot injuries. Additionally, gait timing and stride length were closer to unloaded walking values when subjects wore the device.

Adding a stiff pathway in parallel to the vertebral column, however, also revealed some less desirable results, such as increased dynamic loads on the lower limbs which were reflected on higher muscle activations in triceps and quadriceps muscle groups. One limitation of the current study which will be addressed in future experiments is the relatively small sample size. Also, providing the Second Spine with tunable stiffness/damping between the upper structure and the hip belt may further enhance the effectiveness of our design. For this reason, the next generation of Second Spine, currently being developed, will be equipped with motors and load cells for active compensation of the dynamic loads.

ACKNOWLEDGMENT

This work was supported by DARPA and the Army Research Laboratory through the Warrior Web program.

REFERENCES

[1] J. Knapik, E. Harman, and K. Reynolds, “Load carriage using packs: A review of physiological, biomechanical and medical aspects,” *Applied ergonomics*, vol. 27, no. 3, pp. 207–216, 1996.

[2] A. White and M. Panjabi, *Clinical biomechanics of the spine*. Lippincott Philadelphia, 1990, vol. 446.

[3] M. Feuerstein, S. Berkowitz, and C. Peck Jr, “Musculoskeletal-related disability in us army personnel: prevalence, gender, and military occupational specialties,” *Journal of occupational and environmental medicine*, vol. 39, no. 1, pp. 68–78, 1997.

[4] C. Devroey, I. Jonkers, A. de Becker, G. Lenaerts, and A. Spaepen, “Evaluation of the effect of backpack load and position during standing and walking using biomechanical, physiological and subjective measures,” *Ergonomics*, vol. 50, no. 5, pp. 728–742, 2007.

[5] Y. Hong, J. Li, and D. Fong, “Effect of prolonged walking with backpack loads on trunk muscle activity and fatigue in children,” *Journal of Electromyography and Kinesiology*, vol. 18, no. 6, pp. 990–996, 2008.

[6] W. Mackenzie, J. Sampath, R. Kruse, and G. Sheir-Neiss, “Backpacks in children,” *Clinical orthopaedics and related research*, vol. 409, pp. 78–84, 2003.

[7] J. Knapik, K. Reynolds, and E. Harman, “Soldier load carriage: historical, physiological, biomechanical, and medical aspects,” *Military medicine*, vol. 169, no. 1, pp. 45–56, 2004.

[8] N. Ramanathan, S. Datta, and M. Gupta, “Biomechanics of various modes of load transport on level ground.” *The Indian Journal of Medical Research*, vol. 60, no. 11, pp. 1702–1710, 1972.

[9] E. A. Harman, P. N. Frykman, J. J. Knapik, and K.-H. Han, “785 backpack vs. front-back pack: Differential effects of load on walking posture,” *Medicine & Science in Sports & Exercise*, vol. 26, no. 5, 1994.

[10] H. Kinoshita, “Effects of different loads and carrying systems on selected biomechanical parameters describing walking gait,” *Ergonomics*, vol. 28, no. 9, pp. 1347–1362, 1985.

[11] S. Legg and A. Mahanty, “Comparison of five modes of carrying a load close to the trunk,” *Ergonomics*, vol. 28, no. 12, pp. 1653–1660, 1985.

[12] R. F. Johnson, J. J. Knapik, and D. J. Merullo, “Symptoms during load carrying: effects of mass and load distribution during a 20-km road march,” *Perceptual and Motor Skills*, vol. 81, no. 1, pp. 331–338, 2013/01/30 1995.

[13] K. Reynolds, J. Kaszuba, R. Mello, and J. Patton, “Prolonged treadmill load carriage: acute injuries and changes in foot anthropometry,” DTIC Document, Tech. Rep., 1990.

[14] D. Bloom and A. Woodhull-McNeal, “Postural adjustments while standing with two types of loaded backpack,” *Ergonomics*, vol. 30, no. 10, pp. 1425–1430, 1987.

[15] M. Holewijn, “Physiological strain due to load carrying,” *European journal of applied physiology and occupational physiology*, vol. 61, no. 3, pp. 237–245, 1990.

[16] J. Bachkosky, M. Andrews, R. Douglass, J. Feigley, L. Felton, F. Fernandez, P. Fratarangelo, A. Johnson-Winegar, R. Kohn, N. Polmar *et al.*, “Lightening the load,” DTIC Document, Tech. Rep., 2007.

[17] G. Tierney, G. Stratton, and M. Tzeng, “Design and optimization of an exospine structure utilizing lightweight composites,” in *in Proc. of the Society for the Advancement of Material and Process Engineering Conference, SAMPE 2012*, 2012.

[18] J. Park, P. Stegall, D. Zanotto, V. Vashista, X. Jin, and S. Agrawal, “Design of the second spine: A secondary pathway to transfer loads from the shoulders to the pelvis,” in *Proceeding of the IDETC/CIE 2013, DETC2013-12795*, 2013.

[19] J. Brimacombe, D. Wilson, A. Hodgson, K. Ho, and C. Anglin, “Effect of calibration method on tekscan sensor accuracy,” *Journal of biomechanical engineering*, vol. 131, p. 034503, 2009.

[20] M. Kadaba, H. Ramakrishnan, and M. Wootten, “Measurement of lower extremity kinematics during level walking.” *Journal of orthopaedic research: official publication of the Orthopaedic Research Society*, vol. 8, no. 3, p. 383, 1990.

[21] H. Hermens, B. Freriks, R. Merletti, D. Stegeman, J. Blok, G. Rau, C. Disselhorst-Klug, and G. Hägg, *European recommendations for surface electromyography*. Roessingh Research and Development The Netherlands, 1999.

[22] J. Zeni, J. Richards, and J. Higginson, “Two simple methods for determining gait events during treadmill and overground walking using kinematic data,” *Gait & posture*, vol. 27, no. 4, pp. 710–714, 2008.

[23] J. H. Zar, *Biostatistical Analysis (5th Edition)*. Upper Saddle River, NJ, USA: Prentice-Hall, Inc., 2007.

[24] A. Bjelle, M. Hagberg, and G. Michaelson, “Occupational and individual factors in acute shoulder-neck disorders among industrial workers.” *British journal of industrial medicine*, vol. 38, no. 4, pp. 356–363, 1981.

[25] E. Morin, “Assessing physiological and biomechanical loading using the portable measurement system and the dynamic biomechanical model,” DRDC Document, Tech. Rep., 2008.

[26] G. Ghori and R. Luckwill, “Responses of the lower limb to load carrying in walking man,” *European journal of applied physiology and occupational physiology*, vol. 54, no. 2, pp. 145–150, 1985.

[27] E. Harman, K. Han, P. Frykman, M. Johnson, F. Russell, and M. Rosenstein, “The effects on gait timing, kinetics, and muscle activity of various loads carried on the back: 774,” *Medicine & Science in Sports & Exercise*, vol. 24, no. 5, p. S129, 1992.

[28] S. Birrell and R. Haslam, “The effect of load distribution within military load carriage systems on the kinetics of human gait,” *Applied ergonomics*, vol. 41, pp. 585–590, 2010.

[29] J. Bobet and R. Norman, “Effects of load placement on back muscle activity in load carriage,” *European journal of applied physiology and occupational physiology*, vol. 53, no. 1, pp. 71–75, 1984.

[30] E. Harman, K. Hoon, P. Frykman, and C. Pandorf, “The effects of backpack weight on the biomechanics of load carriage,” DTIC Document, Tech. Rep., 2000.

[31] D. A. Winter, *The biomechanics and motor control of human gait: normal, elderly and pathological*. Univarsity of Waterloo Press, 1991.

[32] K. Han, E. Harman, P. Frykman, M. Johnson, F. Russell, and M. Rosenstein, “Load carriage: the effects of walking speed on gait timing, kinetics, and muscle activity: 773,” *Medicine & Science in Sports & Exercise*, vol. 24, no. 5, p. S129, 1992.

[33] A. Voloshin and J. Wosk, “An in vivo study of low back pain and shock absorption in the human locomotor system,” *Journal of biomechanics*, vol. 15, no. 1, pp. 21–27, 1982.

[34] P. Helliwell, J. Smeathers, and V. Wright, “Shock absorption by the spinal column in normals and in ankylosing spondylitis,” *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 203, no. 4, pp. 187–190, 1989.

[35] R. Norman and D. Winter, “The utility of combining emg and mechanical work rate data in load carriage studies,” in *Proceedings of the 4th Congress of the International Society of Electrophysiological Kinesiology, Boston, MA*, 1979.

[36] J. Bobet and W. Robert, “Use of the average electromyogram in design evaluation investigation of a whole-body task,” *Ergonomics*, vol. 25, no. 12, pp. 1155–1163, 1982.