Lower Extremity Exoskeleton Reduces Back Forces in Lifting

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ABSTRACT
We propose a lower extremity exoskeleton device which adds a passive extensor moment (restoring moment) about the hips during squat lifting, thus reducing forces on the lower back by reducing the required extensor muscle force. Video sequences were recorded of normal speed sagittal squat lifting 44.5 N (10 lb) and 133.5 N (30 lb) packages for marker tracking. Calculations suggested that the device reduces maximum spine compressive forces by approximately 1300 N. Surface electromyography (EMG) was performed on 6 subjects supporting 44.5 N (10 lb) and 133.5 N (30 lb) packages in the static squat posture. With the device, back muscles demonstrated a 54% reduction in muscle activity. This exoskeleton device includes features not available on other devices including highly adjustable moment profile and elimination of high contact stress in the lower extremities by connecting directly with the ground.

INTRODUCTION
In the United States, approximately 2 363 960 people are employed under the designation of Manual Freight, Stock, and Material Movers (Bureau of Labor Statistics) [1]. Among this group, approximately 270 890 reported cases of lower back disorder were severe enough to cause at least one day of missed work in 2006 [2]. Back strain composes 30% of all workplace injuries, the majority of these cases being caused by manual material handling (MMH) [3]. The issue of injury reduction has been addressed through, administration and control solutions, and through engineering solutions. Administrative solutions include training and regulation. Engineering solutions include workplace redesign as well as new devices including powered-lift tables, overhead lift assist systems, and worn devices. Existing lift assist systems are unacceptably slow for many tasks thus reducing productivity. Many warehouse MMH tasks involve movement about a warehouse and activity in areas with limited overhead access, reducing the usefulness of overhead devices to specialized and relatively stationary situations. Even as heavy lifting in the workplace becomes less prevalent, instances of workplace low back injury continue to rise [4] indicating that absolute maximum package weight is not the only consideration.

The National Institute of Occupational Safety and Health (NIOSH) publishes findings and issues guidelines for safe lifting which are enforced by it's parallel agency focused on regulation, the Occupational Safety and Health Administration (OSHA) [9] [10]. The NIOSH lift equation (1981) and revised NIOSH lift equation (1993) recommend maximum lifting loads for repetitive lifting by healthy workers [6], capped at 223 N (51 lb), and reduced based on a number of risk factors [6]. NIOSH recommends a maximum spinal compression of 3400 N during lifting, called the action limit. Note, during bending or squatting, a moment will be generated about the hips (axis of rotation) due to the weight of the body superior to the hips and any load being lifted. The body must generate an equal and opposite moment to prevent toppling forward. Moving superiorly from the hips, there is less mass to support, thus the required restoring moment decreases. 85 to 95 percent of low back injuries occur at the L5/S1 and L4/L5 discs, and are evenly split between the two. The L5/S1 disc (between the L5 spinal vertebra and the Sacrum) experiences the greatest moment during incline and we will focus on this region [14].

Anatomy
The lumbar region contains many passive structures (bone, ligament, and disc) and active structures (muscles). Ligaments important to stabilization of the spinal region include the anterior and posterior longitudinal ligaments around the vertebral body and the disc, the ligamentum flava between adjacent vertebral laminae, and the interspinous and supraspenous ligaments connecting the spinous process of adjacent vertebrae [11]. Muscles of the low back area include the erector spinae group, traveling the length of the spine oriented in a superior/inferior

direction, the serratus posterior inferior, which originates in vertebrae and insert into the ribs. The thoracolumbar fascia composes a superficial layer, and is believed to play an important role in trunk rotation and stabilization of the lower back. Under the erector spinae are the semispinalis, multifidi, and rotators. Included in some biomechanical back models (especially for those studying the L3/L4 disc) are the left and right latissimus dorsi muscles [12]. In viewing a transverse cross section of the torso at the L5/S1 disc from the superior view, other muscles encountered are the rectus abdominus and the internal and external obliques, left and right. Also sometimes considered is the interabdominal pressure (IAP), or pressure in the abdomen caused by the diaphragm and surrounding structures, believed to create a restoring force during extension of the low back in some situations [12 - 14].

Back forces and biomechanical back/torso models

Major factors contributing to spinal compressive force are package weight, subject posture, and distance of forward reach [15]. Lift speed (static, low speed isokinetic, free dynamic, or high speed) [13] [17] [18] has a great effect on the relative magnitudes of muscle coactivation and IAP. Another variable in spinal compression is asymmetric lifting (lateral bending, twisting, and asymmetric loading such as one hand lifting or side reaching) [8] [19] [20]. Ligaments have shown little effect on spine force in typical MMH postures, as they are understood to stabilize the spine in extreme postures such as slip and fall situations [19]. Most biomechanical models can be placed into one of two groups, optimization based and electromyography-assisted models. Optimization based models utilize the concept that the nervous system will minimize a criterion (e.g. muscle stress, spinal compression) and implement an objective function to accomplish this. This technique allows the analysis of a statically indeterminate musculoskeletal system knowing only external factors (basic anthropometry, posture, velocity, acceleration, external load), but do not include muscle coactivation [24-26]. Electromyography-assisted models use muscle activity measurements to describe the state of trunk muscles, thus determining the loading of the spine [12] [15]. Electromyography (EMG) is a method of recording a muscle's activation signal [16], discussed later. Popular electromyography-assisted models include the Marras and Sommerich model [12] [15] and the Granata and Marras vector model [27]. These Electromyography assisted models are the most accurate for determining muscle forces [18], but require extensive instrumentation, detailed anthropometry data (cross section area of each muscle), and are not predictive (each motion must be recorded and can not be extrapolated from other motions).

Worn devices

Several devices currently exist to reduce back forces. Back belts are widely used in the workplace but their effectiveness is questioned [28] [9] and OSHA does not recommended them as personal protective equipment [10]. Several devices are currently being studied at the research level though they are not currently available to the retail market. The Happyback [29] device provides a flexible fiberglass frame to which a chest harness, belt, and leg extensions attach [30]. A second device, the Bendezy, consists of a dorsal member, a bracing assembly, and resilient elements. Upon bending, the weight of the wearer is at least partially borne by the resilient element [31]. A third device, Bending Non-Demand Return (BNDR) consists of a portable spring leveraged device worn via a belt on the hips allowing a person to engage in stoop postures (straight legs and high hip flexion, as is common in crop harvesting) with less exertion and less stress on the vertebral joints and surrounding muscles [32] [33]. A fourth device, the Personal Lift-Assist Device (PLAD), employs resilient members (elastic or tension springs) to stretch as the wearer stoops, supporting at least some of the weight of the upper body [34]. These devices were designed specifically for stoop rather than squat postures (considerable knee flexion, moderate back incline). The devices are specifically designed for static postures, except PLAD which is intended for both sustained stoop and repeated stoop-stand tasks. All of these devices attach to the lower extremities of the user (thigh, shin, ankle, or knee) as anchor points for the devices, and none connects directly to the ground. Laboratory studies show that for sustained stoop postures, all four devices significantly reduce tension in the muscles of the lower back, though user comfort is still a concern [34 - 36].

The purpose of this work is to design, build, and test a worn lower extremity exoskeleton device which reduces forces in the low back during manual material handling tasks and serves as a research platform to investigate lift assisting and torso moment restoring. Further, we aim to study the device and its interface with the wearer, specifically targeting comfort and usability issues and how those balance with maximum effectiveness at reducing low back forces. We also aim to study the human reaction when using the device, specifically determining forces in the body and how they are affected by the device. Our hypothesis is that the device reduces erector spinae muscle loading during squat lifting tasks, and that this correlates to reduced forces in the L5/S1 region of the lower back.

METHODS

A biomechancial model and lift tasks were chosen which balance similarity to real world MMH tasks with repeatability and similarity to published studies. Many prior studies have investigated two-dimensional (sagittal plane) isometric postures, a restriction which increases accuracy of muscle force calculations but reduces similarity to realistic applications [8]. IAP is believed by some not to be a large factor in static or low speed lifting [14] and may even be a byproduct of muscle activity [14] [17] [37]. Granata and Marras have shown that as speed increases, antagonistic activation and IAP, increase [13]. In their study, speeds to 90 degrees per second were studied (acceleration not given), where typical velocities in MMH tasks are found to be 10 to 30 degrees per second [12]. With these considerations in mind, we chose to study static and normal speed (that which is comfortable to the subject) sagittal lifting of 44.5 N (10 lb) and 133.5 N (30 lb) packages to evaluate the effectiveness of the device. In addition to repeatability of lifts, and the quantity of similar published studies, these tasks were similar to those in a study by Takahashi et al [23], including direct spinal pressure measurement through surgical implantation of an intradiscal pressure (IDP) transducer. Results showed correlation between applied moment (torso weight plus applied load), muscle activation, and spine force. Results demonstrated that while measured disc pressure was higher than expected, the offset between theoretical and measured pressures during lifting tasks was predictable. Because the goal of our work was to determine the effectiveness of our device, we sought a reduction in back forces rather than the total force magnitude. So although spine forces were most likely higher than calculated values, we were able to estimate force reduction caused by the device.

The reduced-force back model

Chaffin and Anderson presented the reduced-force back model [14] for sagittal lifting, in which coactivation and IAP are not considered. In the sagittal case, many terms in the three-dimensional low-back model from Schultz and Anderson [38] [14] canceled, and the equation in the more involved 3D model reduced to the reduced-force back model. Also, optimization models which sought to minimize overall force demonstrated the same term canceling effect and became equivalent to the reduced-force back model for sagittal lifting. With all of these considerations, it was decided that sagittal lifting and the reduced-force back model discussed below struck the best balance between accuracy, repeatability, correlation with a real MMH tasks, and correlation with published experiments. For sagittal lifting, the reduced force back model becomes an optimization method, as it reduces non-extensor muscle forces to their minimum, zero.

External loads to be considered in calculating back forces for static squatting (dynamic effects discussed below) were the weight of the upper body (body) and the weight of the object being lifted (package). Body is defined as the torso superior to the L5/S1 disc, head, neck, and upper extremities. These forces caused direct spinal compression, and in non-neutral postures caused a moment about the L5/S1 disc as a function of the weight and the lateral distance from the disc as shown in fig. 1A and given in Eq. (1).

M_{APP} Is the total applied moment (from body weight and external weights). F_{BODY} and F_{PKG} are the weights of the body and package, respectively. L_{BODY} and L_{PKG} are the lateral distances from the L5/S1 disc to the body and package CG's respectively (fig. 1A). To maintain static equilibrium ($\Sigma M = 0$, $\Sigma Fx = 0$, $\Sigma Fy = 0$), an equal and opposite moment about the spine was generated, through tension in the erector spinae muscles (Fig. 1B). Distance from the erector spinae to the center of the L5/S1 disc (distance D in Fig. 1B) has been found to be 0.065m (2.6 in) [14]. Anthropometric values including location of L5/S1 and CG were for the 50th percentile male [14]. Muscle and spine forces were solved as a function of bend angle α from vertical (Fig. 1A), Eq. (2) and Eq. (3).

$$M_{APP} = [F_{BODY} \text{ x } L_{BODY}] + [F_{PKG} \text{ x } L_{PKG}] \quad (1)$$
$$F_M = [M_{APP}] / 0.065\text{m} \quad (2)$$
$$F_S = F_M + [F_{BODY} + F_{PKG}] \text{ x } \cos(\alpha) \quad (3)$$

Fm was muscle force (tension), and Fs was spine force (compression). For force equilibrium, a compressive force in the spine is created equal to erector spinae muscle tension plus the cosine portion of the body and package weights (Fig. 1B), (Eq. 3). For illustration, Fig. 1C shows theoretical muscle and spine force as a function of back incline angle, alpha, during sagittal squat lifting a 133.5 N (30 lb) package. To reduce the force on the low back, a restoring moment was applied to the torso, sketched as a lower extremity exoskeleton in Fig. 1D. Equations (1) to (3) were modified to yield Eqs. (4) to (6) and include a term for the restoring moment due to the device.

$$\text{Total Moment} = M_{APP} - M_{DEV} \quad (4)$$
$$F_M = [M_{APP} - M_{DEV}] / 0.065\text{m} \quad (5)$$
$$F_S = F_M + [F_{BODY} + F_{PKG}] \text{ x } \cos(\alpha) \quad (6)$$

M_{DEV} was the restoring moment due to the device. This restoring moment directly reduced muscle tension (Eq. 5), which in turn reduced spine compressive forces (Eq. 6). All lifts were performed at normal speed, as chosen by the subject, to most accurately reflect the workplace lifting condition.

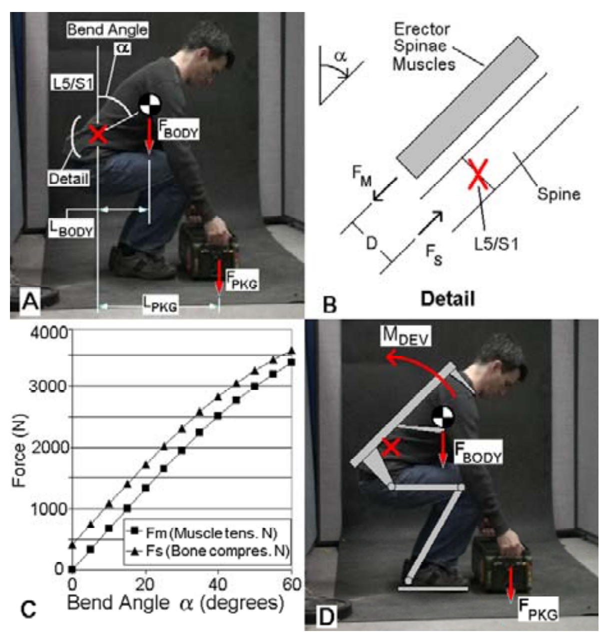

FIGURE1. MOMENTS AND FORCES ON THE LOW BACK. A. FORCES, DISTANCES, AND BEND ANGLE DURING SQUAT LIFTING. B. INTERNAL FORCES TO MAINTAIN EQUILIBRIUM. C. CALCULATED FORCES ON THE L5/S1 DISC AS A FUNCTION OF BEND ANGLE LIFTING A 133.5 N (30 LB) PACKAGE. D. DEVICE WITH RESTORING MOMENT M_{DEV}.

The device

A passive wearable moment restoring device was designed and built from the lower extremity exoskeleton in the UC Berkeley human engineering laboratory. The device delivered a restoring moment to a user thus reducing low-back muscle tension and L5/S1 compression during lifting. The device served as a research platform allowing us to determine an appropriate balance between maximum restoring moment and wearer comfort, and to study the interface between human user and moment restoring device. The device was attached to the wearer via chest and shoulder harness, hip straps, and thigh straps. The device connected to the ground with the wearer stepping into foot bindings (similar to snowshoe or snowboard bindings) rather than terminating at the thigh or ankle, thereby eliminating high contact stresses on the wearer's lower extremities. Restoring moment was generated at the hip (axis of rotation during squat) via a spring cable mechanism absorbing energy during squat, and returning the energy when the wearer returned to neutral. Cable wrapped around the star-cam during squat, extending the springs, and creating a moment, F x L. Screws on the star-cam could be moved nearer or farther from the hip pivot to modify restoring moment. Tensioner was used to adjust onset angle of restoring moment Fig. (2). Restoring moment was delivered via a force on the wearer's upper torso through the chest/shoulder harness opposite the direction of bend (counterclockwise about the hips in Fig. 1D). Equilibrium counter-forces were exerted through the mechanism and to the ground with minor forces on the back, and at the thigh straps. Mounted at the hip was a "Star-cam". This provided an initial free zone (no restoring moment) allowing the user to perform basic tasks without onset of the restoring moment (legs were free during walking or stepping over objects). Restoring moment was calculated using the known spring rate and moment arm characteristics of the star-cam. These calculations were verified through pull-tests of the device alone (no wearer) using a pull spring scale, and correlated to within 10% of calculated values.

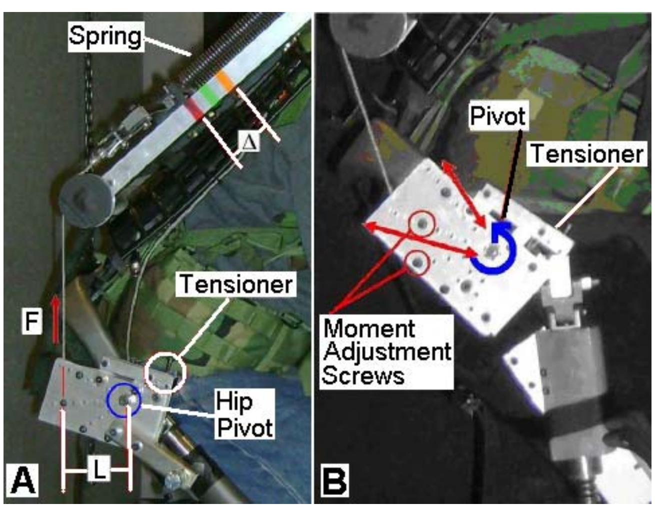

FIGURE 2. RESTORING MOMENT MECHANISM. A. STAR-CAM. B. STAR-CAM.

The experiment

The study consisted of six subjects, five male, one female, mean height 1.75m (std. dev. 0.06 m), mean mass 67.7 kg (std. dev. 7.2 kg), and median age 27.7 year (std. dev. 6.0 year). All were in good health, and none had prior back injury or history of back pain. Subjects were given an orientation session and were informed that the test was voluntary and to stop immediately if they experienced pain or discomfort and to rest when needed. Subjects were fit to the device and asked to walk and squat, initially while holding a support rail. Device fit was adjusted, and the procedure repeated until the device fit comfortably. Restoring moment was gradually increased until it was considered uncomfortable then was reduced to determine the maximum comfortable level. Restoring moment was modified over portions of the bend cycle until repeated squats were completed with the subject satisfied that further modifications would not increase comfort. The subject removed the device and was asked to comment on device comfort. Final star-cam settings were recorded.

Marker tracking

Video analysis was performed in order to evaluate the effectiveness of the device in dynamic lifting. Due to logistics, it was not possible to record and track marker sequences for all subjects, so sequences were recorded and tracked for one subject, and individual frames from video sequences of other subjects were cross-analyzed to compare with tracked marker sequences. Retro-flective markers (2.5 cm diameter) were adhered to the device, and the package. Marker locations were tracked using custom code written for Matlab mathematical software (The Mathworks, Inc. Natick MA.).Video sequences were recorded normal to the sagittal plane as the subject squat-lifted 44.5N (10lb) and 133.5 N (30 lb) packages. Package dimensions, 0.55 m (22 in) wide at handles, with steel rod handles 0.13 m (5 in) in length and 0.15 m (6 in) from the ground. The subject performed each lift at normal speed, as chosen by the subject. Each sequence consisted of standing in a vertical neutral posture, squatting to the package, lifting the package until subject stood erect, lowering the package to the ground, and returning (without the package) to the neutral posture. Motion in Z (toward the camera), such as spreading the arms slightly to grasp the handles, was not included. Markers were tracked and L5/S1 disc, upper body center of gravity (CG), and the hip joint were calculated for each frame (Fig. 3).

Forces for the pseudo-static case were calculated (posture in each frame was analyzed and static free body forces were calculated using Eqs. 1 through 6). From frame rate and parker positions, velocity and acceleration were determined and dynamic forces were calculated between frames. As the frame rate (29.97 frames per second) was high relative to the marker velocities, we approximated $d/dt \approx \Delta/\Delta t$ to calculate velocity and acceleration:

$$V_x = dX/dt \quad (7)$$
$$V_y = dY/dt \quad (8)$$

$$A_{x\,pkg} = d^2X_{pkg} / dt^2 \quad (9)$$
$$A_{y\,pkg} = d^2Y_{pkg} / dt^2 \quad (10)$$
$$A_{x\,bdy} = d^2X_{bdy} / dt^2 \quad (11)$$
$$A_{y\,bdy} = d^2Y_{bdy} / dt^2 \quad (12)$$

V and A were velocity and acceleration in X and Y indicated by subscripts, accelerations corresponding to the package or body as indicated in the subscript. This allowed us to calculate dynamic components (X and Y) of the force equations. The forces in Eq. (1) become:

$$F_{x\,bdy} = [m_{bdy} \text{ x } A_{x\,bdy}] \quad (13)$$
$$F_{y\,pkg} = [m_{pkg} \text{ x } g] + [m_{pkg} \text{ x } A_{y\,pkg}] \quad (14)$$
$$F_{x\,pkg} = [m_{pkg} \text{ x } A_{x\,pkg}] \quad (15)$$
$$F_{y\,bdy} = [m_{bdy} \text{ x } g] + [m_{bdy} \text{ x } A_{y\,bdy}] \quad (16)$$

F terms are force in X or Y, due to the mass of the package or body as indicated by subscripts. The term g is the gravitational constant, m_{bdy} is the mass of the body, m_{pkg} is the mass of the package. Expanding the Y forces to the dynamic case and adding the dynamic X forces to the analysis. Applied moment was then summed, as in Eq. (1). Equations (4) to (6) were computed with these new values, expanding analysis to the dynamic case.

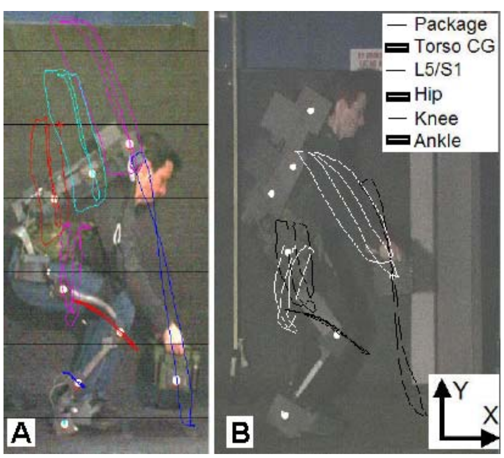

FIGURE 3. MOTION TRACKING. A. MARKERS TRACKED DURING LIFT. B. PATHS OF KEY POINTS CALCULATED.

EMG

EMG data was taken using a Bipolar surface EMG recording system (BlackRoseTechnology, Oakland, CA. USA). The signal was rectified, low pass filtered, and output to an Instrunet model 100B Analog/Digital Input/Output system (Instrunet, Somerville, MA. USA.) at 64 points per second. As only static tests were performed, the signal was averaged over the full five second posture. Ten millimeter diameter silver electrodes were prepared with electrolytic gel, and adhered in place over subjects' erector spinae muscles, approximately 3 cm apart, over the muscle belly near the L5/S1 disc. To calibrate, subjects squatted to "lift" a stationary horizontal rail, 200 mm above the ground, and exerted maximum effort to raise the stationary bar for five seconds as maximum voluntary contraction (MVC). Calibration was carried out twice for each subject, and the higher average response was used as MVC. EMG data is reported as percent of MVC. Subjects squatted to lift a 44.5 N (10 lb) package slightly but no more than 10 cm (4 in) off the ground (same package used in marker tracking trials), and maintained that position for five seconds then returned to neutral, then repeated the task. Consistent posture was maintained for each trial, including tucked knees, arms outstretched forward holding the package, back inclined (α) 45 to 50 degrees. The task was repeated. Subjects repeated the static squat lift two more times with a package of 133.5 N (30 lb). Upon completion, subjects donned the moment restoring device and repeated the procedure. In total, each subject completed 2 trials at 2 weights in 2 lift states, totaling 8 lifts per subject.

RESULTS

A balance was reached between maximum restoring moment and maximum user comfort. It was found that proper fit of the device itself was critical, and much adjustment was necessary prior to any restoring moment comfort evaluation. Comfort tests and interviews with six subjects indicated that the device was most comfortable if it included an initial free zone, providing no restoring moment, which allowed users to walk freely and step over small objects. Subjects felt uncomfortable experiencing a large restoring moment during moderate squat, but in maximum squat, a substantial restoring moment was considered acceptable. Greater restoring moment at higher flexion angle reduced back forces where the reduction was most needed. The force results are shown in Fig. 4. After subject comfort testing was complete, one star-cam setting was chosen based on comfort trials and used for all testing. Final configurations chosen by subjects varied only slightly, the primary difference being torque onset angle. No subject requested moment profile modification. One subject (with the lowest body mass) requested a lower overall spring rate in both initial comfort and follow up EMG testing. The number of springs was reduced from six to four, reducing overall restoring moment by one third for that subject, but maintaining restoring moment profile.

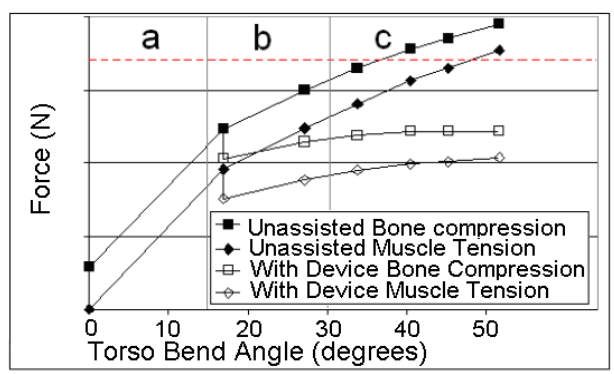

FIGURE 4. RESTORING MOMENT VS BACK FORCES. a INITIAL FREE ZONE. b. SMALL RESTORING MOMENT. c. LARGE RESTORING MOMENT

Marker tracking indicates that erector spinae muscle tension and L5/S1 disc compression were reduced when the exoskeleton device was used, especially at greater bend angles due to nonlinear moment profile as determined previously. Lift sequence consists of four motions (Fig. 5A) with corresponding portions of the force curve (Fig. 5B).

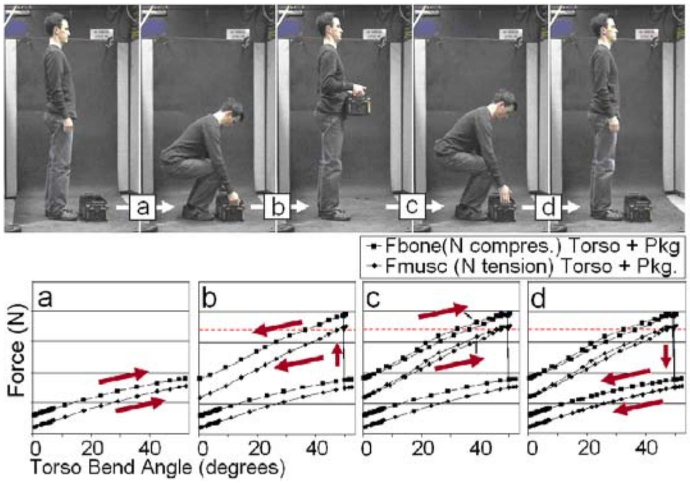

FIGURE 5. SEGMENTS OF LIFT. TOP, LIFT DIVIDED INTO SUB-MOTIONS. BOTTOM, FORCES FOR EACH SUB-MOTION. a. SQUAT TO THE PACKAGE. b. FORCES RISE AS PACKAGE IS LIFTED THEN DECLINE AS TORSO BEND ANGLE IS REDUCED. c. FORCES INCREASE WITH INCREASED BEND ANGLE d. PACKAGE IS RELEASED AND SUBJECT RETURNS TO NEUTRAL POSITION.

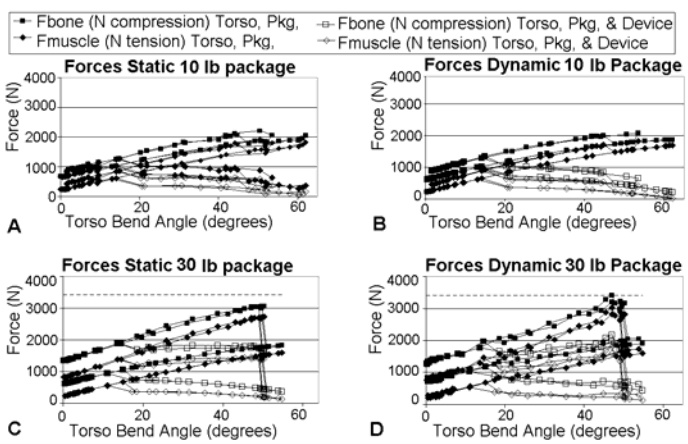

FIGURE 6. FORCES WITH AND WITHOUT DEVICE. LOW BACK FORCE VERSUS BEND ANGLE, STATIC AND DYNAMIC FOR 44.5 N (10 LB) AND 133.5 N (30 LB) PACKAGES.

Back forces are shown in Fig. 6A over the course of the squat lift. Maximum forces for each case are given in Tab. 1. Dynamic analysis shows a 12% increase in maximum force with a 133.5 N (30 lb) package, and 6% with a 44.5 N (10 lb) package. Overall force reduction with the device ranges from 1237 N to 1422 N. Takahashi showed that muscle forces are underestimated, but the difference between forces is accurate [23]. Marker tracking was performed on only one subject, but video sequences of all subjects were compared as they performed the dynamic lift tasks over three trials. One frame from each trial was selected, just after the package was lifted from the ground. This was believed to be the worst loading as both L_{PKG} and L_{BODY} are their largest. Postural data was taken from each of the chosen frames and compared to the marker tracked frame, to calculate restoring moment and corresponding force reduction. Comparing postural data among the subjects indicated that force reduction in the static case was 1412 N average (std dev 85) when lifting the 133.5 N package, and 1515 N (std dev 77) for the 44.5 N (10 lb) package. As this was single frame data, it can only be used to find static properties.

TABLE 1. BACK FORCES, STATIC, DYNAMIC, WITH AND WITHOUT DEVICE.

	Static		Dynamic	
	No Device	With Device	No Device	With Device
44.5 N Pkg.	2075 N	653 N	2211 N	878 N
133.5 N Pkg.	3052 N	1741 N	3420 N	2183 N

Normalized static EMG tests for six subjects showed statistically significant drops in erector spinae muscles when using the moment reduction device. Table 2 shows results for 44.5 N (10 lb) and 133.5 N (30 lb) lifts.

TABLE 2. RESULTS OF STATIC EMG TESTING

	No Device		With Device		
Pkg Wt.	Mean (% MVC)	Std. Dev.	Mean (% MVC)	Std. Dev.	% reduc.
44.5 N Pkg.	22.7	6.3	12.6	8.5	44%
133.5 N Pkg.	39.0	8.5	18.1	8.5	54%

DISCUSSION

Our hypothesis was that the novel device reduced erector spinae muscle loading during squat lifting tasks, and that this correlates to reduced forces in the L5/S1 region of the low back. Static and dynamic analysis of markers tracking, as well as static EMG testing showed that erector spinae muscle force was greatly reduced in the tested configuration. Other goals of this work were to study human-device interface (comfort) and device effectiveness at reducing spine and muscle forces. The device performed as designed, generating a restoring moment at the hips, reducing erector spinae muscle activation among six subjects. Analysis shows, and is supported by other studies [23], that reduced erector spinae tension directly reduces spine compression. The moment restoring design allowed for real-time moment modification during comfort testing, yielding a more comfortable and effective device. It is believed that a further testing on restoring moment profile will narrow the required window of adjustability. For that reason, once comfort testing yielded a suitable candidate, no further moment profile adjustments were made.

Back forces were measured indirectly through marker tracking, and muscle activity levels were measured directly using static surface EMG, resulting in a 54% reduction in erector spinae activity when lifting a 133.5 N (30 lb) package. Calculations show that the reduced erector spinae tension yields a reduced spinal compression, which was confirmed by Takahashi in 2006 using direct IDP measurements in static postures and in motions very similar to the ones studied here [23]. Limiting tasks to sagittal movement allowed us to compare our chosen reduced-force biomechanical model [14] and resulting forces to a wide range of previous test data, including comparing our device to similar tests with other devices [35] [36]. Barrett and Fathallah tested most available devices, using static stooped postures, finding EMG reduction [36] ranging from 21% to 31%. Other trunk muscles were also tested, yielding little activation, thus little change when devices were added. This agrees with work by Marras [12] and optimization models [24] [26] indicating that static sagittal postures demonstrate little coactivation from other trunk muscles. Our device includes several features not present in the other designs. As our device delivers a restoring moment to the user based on the thigh-torso angle, the wearer can squat or stoop. Our device contains a high degree of restoring moment profile adjustability, an area not covered by other devices. Wearers of each of the other devices report discomfort due to high contact stress in the lower extremities where the devices are anchored [36] [39]. Our unique use of exoskeleton technology creates a passive device with circumvents this issue by connecting directly to the ground, using it as the lower anchor point, and allowing the device to support itself.

The decision to limit our scope to sagittal lifting and reduce our analysis to the reduced force model was beneficial in that the postures and movements studied were well understood, and there is a wealth of correlating data and corresponding biomechanical models available for reference [14] [23]. The decision to limit EMG testing to the static case was made in order to ensure high quality data in a known configuration, and to improve comparison with other devices, each of which was tested under static sagittal EMG conditions [34] [36]. Though narrowing the scope was appropriate for this first phase (design, build, tune, and initial test), it would be very desirable to expand the scope of testing and remove some of the restrictions for the next level of device evaluation. Future work will include expanding EMG testing into dynamic tasks, comparing erector spinae and other trunk muscle activation levels while squat lifting packages at a normal pace. Next, testing will be expanded into asymmetric tasks, using tasks which can be repeated with little variation, but which represent realistic MMH tasks. The current reduced force back model does not accommodate the forces involved in such tasks, so a new biomechanical model must be chosen.

Limitations of this study include small sample size, especially of female subjects, study of erector spinae muscles only, and limiting the EMG study to static sagittal lifting.

CONCLUSION

Results from all three sets of experiments (static, dynamic, and EMG) show that the device substantially, and statistically significantly, reduces erector spinae muscle forces during the squat lifting task. It has been shown analytically and experimentally that this correlates to a reduction in spine compression forces. The device has been shown to be effective in reducing low back forces during normal speed sagittal squat lifting of moderate weight packages. These results are encouraging and are being used as a framework upon which to build a wider range of experiments.

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REFERENCES

[1] Bureau of Labor Statistics, Occupational Employment and Wages 2005: http://www.bls.gov/oes/current/oes537062.htm

[2] Nonfatal Occupational Injury and Illness requiring days away from work in 2005, Bureau of labor statistics, USDL 06-1982. Nov. 17, 2006. http://www.bls.gov/news.release/pdf/osh2.pdf

[3] Marras WS, Granata KP, Davis KG Allread WG, Jorgensen MJ., 1999. Effects of box feature on spine loading during warehouse order selecting. *Ergonomics,* 42(7), pp.980-995.

[4] Marras WS., 2000. Occupational low back disorder causation and control. *Ergonomics,*43(7), pp. 880-902.

[5] Marras WS, Fine LJ, Ferguson SA, Waters TR., 1999. The effectiveness of commonly used lifting assessment methods to intensify industrial jobs associated with elevated risk of lo-back disorders. *Ergonomics* 42(1), pp. 229-245.

[6] NIOSH Publication No 94-110:*Applications Manual for the Revised NIOSH Lifting Equation*, 1994.

[7] Chengalur SN, Rodgers S, Bernard TE. *Kodak's ergonomic design for people at work*. (2nd ed.). Hoboken, NJ, John Wiley & Sons, 2004.

[8] Marras WS. Lavender SA. Leurgans SE. Fathallah FA. Ferguson SA. Alread WG. Rajulu SL., 1995. Biomechanical risk factors for occupationally related low back disorders. *Ergonomics,* 38 (2), pp. 377-410.

[9] NIOSH Publication No. 1994-122: *Workplace Use of Back Belts Review and Recommendation,*. May 1994.

[10] OSHA Standard 1910.900. *Final Ergonomics Program Standard*, Regulatory Text, November 2000.

[11] Ebraheim NA. Hassan A. Lee M. Xu R., 2004. Functional anatomy of the Lumbar Spine. *Seminars in Pain Medicine* 2004, 2, pp.131-137.

[12] Marras WS. Sommerich C. A., 1991. Three-Dimensional Motion Model of Loads on the Lumbar Spine: I. Model Structure. *Human Factors,* 33(2), pp. 123-137.

[13] Granata KP. Marras WS., 1995. The influence of Trunk Muscle Coactivity on Dynamic Spinal Loads. *Spine*, 20(8), pp 913-919.

[14] Chaffin DB, Andersson G. *Occupational Biomechanics*. 4th ed. New York: John Wiley & Sons; 2006.

[15] Marras WS. Sommerich C. A., 1991 Three-Dimensional Motion Model of Loads on the Lumbar Spine: II. Model Validation. *Human Factors,* 33(2), pp. 139-149.

[16] De Luca CJ., 1997. The Use of Surface Electromyography in Biomechanics. *Journal of Applied Biomechanics*, 13, pp. 135-163.

[17] Schultz A. Andersson G. Ortengre R. Haderspeck K/ Nachemson A., 1982. Loads on the lumbar spine Validation of a Biomechanical Analysis by Measurements of Intradiscal Pressures and Myoelectric Signals. *Journal of Bone and Joint Surgery*, 64-A(5), June.

[18] Granata KP. Marras WS., 1995. AN EMG-ASSISTED MODEL OF TRUNK LOADING DURING FREE-DYNAMIC LIFTING. *Journal of Biomechanics*. 28(11), pp 1309-1317.

[19] Marras WS. Granata KP., 1997. SPINE LOADING DURING TRUNK LATERAL BENDING MOTIONS. *Journal of Biomechanics*, 7, pp. 697-703.

[20] Wilke HJ. Neef P. Hinz B. Seidel H. Claes L., 2001. Intradiscal pressure together with anthropometric data – a data set for the validation of models. *Clinical Biomechanics* 16(Suplement 1) s111-s126.

[21] Nachemson A. Morris JM., 1964. In Vivo Measurement of Intradiscal Pressure. *The Journal of Bone and Joint Surgery* 46-A(5), July.

[22] Wilke HJ. Neef P. Caimi M. Hoogland T. Claes LE., 1999. New In Vivo Measurements of Pressures in the Intervertebral Disc in Daily Life. *Spine*, 24(8), pp 755-762.

[23] Takahashi I. Kikuchi S. Sato K. Sato N., 2006. Mechanical Loading of the Lumbar Spine During Forward Bending Motion of the Trunk-A Biomechanical Study. *Spine*, 31(1), pp 18-23.

[24] Hughes RE. Chaffin DB. Lavender SA. Andersson GBJ., 1994. Evaluation of Muscle Force Prediction Models of the Lumbar Trunk Using Surface Electromyography. *Journal of Orthopaedic Research*, 12(5).

[25] Stokes IAF. Gardner-Morse M., 1999. Quantitative anatomy of the lumbar musculature. *Journal of Biomechanics*. 32, pp. 311-316.

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

[27] Marras WS. Granata KP. A Biomechanical Asessment and Model of Axial Twisting in the Thoracolumbar Spine. *SPINE,* 20(13), pp 1440-1451.

[28] Department of Defense INSTRUCTION NUMBER 6055.1 August 19, 1998. SUBJECT: *DoD Safety and Occupational Health* (SOH) Program.

[29] ErgoAg Company, P.O. Box 1087, Aptos CA. 95001

[30] Roberts B, Inventor. Back Mounted Mobile Back Support Device. US Patent 5,951,591. 1999 Sep. 14.

[31] Mitchell TJ, inventor. Upper Body Support. US Patent 6,436,065. 2002 Aug. 20.

[32] Deamer RM, Deamer DW, Stoop Laborer's Body Support Having Hinge with Adjustable Spring Biasing. US Patent 4,829,989. 1989 May 16.

[33] Anderson R, Deamer RM, Inventors. Stoop Labor Assist Device. US Patent 5,176,622. 1993 Jan 5.

[34] Abdoli-Eramaki M, inventor. Lift Assist Device and Method. Canadian Patent 2,547,270. 2005 June 23

[35] M . Abdoli-E , M . Agnew , J . Stevenson. An on-body personal lift augmentation device (PLAD) reduces EMG amplitude of erector spinae during lifting tasks. *Clinical Biomechanics* , 21(5), pp. 456 – 465.

[36] Barrett, A. Fathallah, F. Evaluation of Four Weight Transfer Devices for Reducing Loads on the Lower Back During Agricultural Stoop Labor. *ASAE Meeting Paper No. 01-8056*. St Joseph, Mich.: ASAE

[37] Davis KG. Marras WS., 2000. The effects of motion on trunk biomechanics. *Clinical Biomechanics,* 15, pp. 703-717.

[38] Schultz A. Andersson G. 1981. Analysis of loads on the lumbar spine. *Spine* 6(1), pp. 76-82.

[39] Abdoli-E M. Stevenson JM., 2008. The effect of an on-body lift assistive device on the lumbar 3D dynamic moments and EMG during asymmetric freestyle lifting. *Clinical Biomechanics* 23, pp. 372-380.