Feasibility of a Biomechanically-Assistive Garment to Reduce Low Back Loading During Leaning and Lifting

Erik P. Lamers, Aaron J. Yang, and Karl E. Zelik

Abstract*—*Goal: **The purpose of this study was: 1) to design and fabricate a biomechanically-assistive garment which was sufficiently lightweight and low-profile to be worn underneath, or as, clothing, and then 2) to perform human subject testing to assess the ability of the garment to offload the low back muscles during leaning and lifting. *Methods:* We designed a prototype garment which acts in parallel with the low back extensor muscles to reduce forces borne by the lumbar musculature. We then tested eight healthy subjects while they performed common leaning and lifting tasks with and without the garment. We recorded muscle activity, body kinematics, and assistive forces. *Results:* The biomechanically-assistive garment offloaded the low back muscles, reducing erector spinae muscle activity by an average of 23–43% during leaning tasks, and 14–16% during lifting tasks. *Conclusion:* Experimental findings in this study support the feasibility of using biomechanically-assistive garments to reduce low back muscle loading, which may help reduce injury risks or fatigue due to high or repetitive forces. *Significance:* Biomechanically-assistive garments may have broad societal appeal as a lightweight, unobtrusive, and cost-effective means to mitigate low back loading in daily life.**

***Index Terms*—Electromyography, exoskeletons, low back pain, smart clothing, spine loading.**

I. Introduction

LOW back pain is a disabling condition experienced by an estimated 60-85% of adults within their lifetime [1]–[3]. Low back pain is the leading cause of limited physical activity [1], the second leading cause of missed work in the U.S. [4], and a significant economic burden. It is estimated to cost >$100 billion per year in the U.S. due to medical expenses and lost worker productivity [5], [6].

The etiology of low back pain is multifactorial, but major risk factors include high (overloading) and/or repetitive (overuse) forces on lumbar tissues (muscles, ligaments, vertebrae and intervertebral discs). Such forces can occur during common daily activities such as leaning and lifting [7]–[9], in both occupational and non-occupational settings. Low back pain is particularly common among individuals who perform heavy lifting [10]–[14]. Static bending and twisting postures, and to a lesser extent prolonged static work postures (e.g., leaning), have also been implicated as potential risk factors for low back pain [15]. Elevated or even moderate loads applied repetitively to the lumbar spine can increase the risk of low back pain [10]–[12], [16], [17], weaken or damage the vertebrae [18]–[20] and cause intervertebral disc degeneration or herniation [21]. Similarly, elevated and repetitive loading of passive (e.g., ligament) or active (muscle) tissues can cause fatigue, discomfort or damage, such as strains [22]–[25]. Reducing loading on lumbar tissues during daily activities could help lower the risk of back injury and resultant pain, or benefit other outcomes related to muscle effort and fatigue.

During tasks such as lifting and leaning, back extensor muscles and ligaments apply the majority of the compression force experienced by the lumbar spine. The lumbar spine experiences a flexion moment during forward leaning due to the weight of the upper-body (F_{HAT}, Fig. 1) and any additional external (e.g., carried) weight (F_W) which act at moment arms (r_{HAT} and r_W) about a given spinal disc (Fig. 1). To prevent the upper-body from falling forward, this flexion (clockwise) moment must be counter-balanced by an extension (counter-clockwise) moment. This extension moment is provided by back extensor muscles (e.g., erector spinae muscles), and passive tissues (e.g., interspinous ligaments, Fig. 1(b)). These muscles and ligaments act at much smaller moment arms than the upper-body center-of-mass and external weight [8], [26]; and therefore produce much larger forces in order to generate the required extension moment [27]. Muscles and ligaments apply forces roughly parallel to the lumbar spine. As a result, they also apply substantial compression forces to the lumbar spine when they are loaded [8] (Fig. 1). It has been shown that extensor muscles and ligaments are responsible for the majority of the

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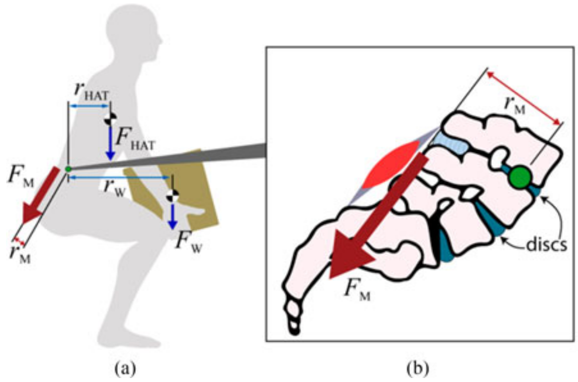

Fig. 1. (a) During tasks such as lifting or leaning, the lumbar (low back) extension moment is the result of large muscle and ligament forces (F_M) acting at short moment arms (r_M) about the spine. This extension moment counterbalances flexion moments from the weight of the head-arms-trunk (F_{HAT}) and carried weight (F_W), which act at larger moment arms (r_{HAT} and r_W). (b) The large muscle and ligament forces constitute the majority of loading experienced by the intervertebral discs. Excessive and/or repetitive loading can lead to degeneration or injury of the lumbar tissues (muscles, ligaments, vertebrae, and intervertebral discs).

compression force experienced by the lumbar spine during tasks such as forward leaning [12], [28], [29]. Reducing these muscle and/or ligament forces may therefore also reduce forces on the spine.

Wearable assistive devices (e.g., exoskeletons, orthoses) provide potential means for reducing lumbar tissue loading, which may then reduce the risk of force-induced low back injury and associated pain. Various wearable assistive devices have recently been developed, which assist via actuators (e.g., [30]–[33]) and/or springs (e.g., [34]–[38]). These devices apply forces to the body in order to provide an assistive extensor moment about the lumbar spine; this offloads the lumbar extensor muscles. Because these devices are worn outside the body they operate with a larger moment arm (often >20 cm [34], [39]) than muscles (6-8 cm [26], [40]) and ligaments (1-5 cm [8]). Therefore, these devices can provide equivalent extensor moments with smaller force magnitudes, resulting in reduced lumbar spine compression forces [8], [39]. Experimental studies indicate that wearable assistive devices can reduce extensor muscle loads by, on average, 9-54% as evidenced indirectly by reductions in muscle activity [34], [35], [37], [41], [42].

The majority of these wearable devices have relatively bulky form-factors, which make them less practical for at-home daily use, or use in other business, social or clinical settings. Often these devices are designed with components that protrude from the lower back to lengthen the extensor moment arm; however, these components can interfere with common daily activities (e.g., sitting, stair ascent/descent, navigating home or work environments). Furthermore, users are generally required to wear these exoskeletal devices conspicuously on top of their clothing, which can be undesirable.

An appealing alternative to these exoskeletal devices may be to adapt form-fitting clothing itself by embedding passive (spring-like), quasi-passive (clutchable spring-like), or active (actuated) structures to assist the low back musculature. We propose that a garment embedded with load-bearing structures may provide similar offloading benefits as exoskeletons but without the burden of bulky components. The key distinguishing feature of our proposed garment-like device is that it could be worn underneath or integrated into clothing; whereas current powered and passive assistive devices are worn on top of clothing. However, it is currently unclear to what degree such a form-fitting garment could offload lumbar tissues. Therefore, the purpose of this initial study was: (i) to design and fabricate a biomechanically-assistive garment which was sufficiently lightweight and low-profile to be worn underneath (or as) clothing and then (ii) to perform human subject testing to assess the ability of the garment to offload the low back during leaning and lifting.

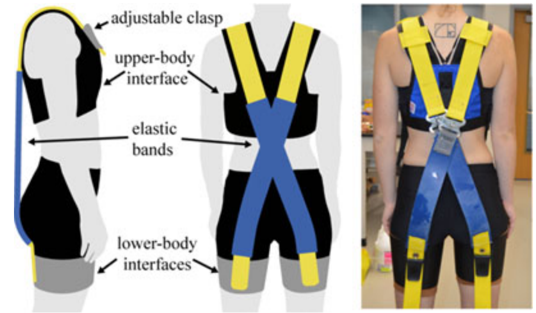

Fig. 2. Biomechanically-assistive garment prototype. The prototype distributes forces over the shoulders and thighs via the upper- and lower-body interfaces. Elastic bands connect these two interfaces and act in parallel with muscles and ligaments to support the lumbar extension moment. Slack length of the elastic bands was set via adjustable clasps.

II. Materials and Methods

A. Prototype Design & Function

We designed and fabricated a passive (unactuated) biomechanically-assistive garment prototype that provides an extension moment about the user's lumbar spine during forward leaning and lifting [43]. The prototype is composed of an upper-body interface (shirt), lower-body interface (shorts), and elastic bands which run along the back, coupling the upper and lower-body interfaces (Fig. 2). The lower-body interface was comprised of thigh sleeves with a high friction elastomer interior and a sturdy fabric exterior, which served as an anchoring point for the elastic bands. The upper and lower interfaces distribute forces across the surface area of the shoulders and thighs, respectively, allowing them to support substantial loading. The elastic bands (6.3 cm × 0.45 cm × 29 cm, ~800 N/m stiffness) were connected to clasps on the shoulders allowing the slack length in the bands to be adjusted. The total weight of the prototype, without testing-related instrumentation, was 2 kg.

As the user leans forward via lumbar flexion and/or hip flexion, the elastic bands stretch, applying tension forces that act roughly parallel to the lumbar extensor muscles and ligaments, and thus generate an extension moment about the lumbar spine (Fig. 3). Because the elastic bands act at a larger moment arm (>9 cm) than biological tissues (~1-8 cm), the biomechanically-assistive garment provides a mechanical

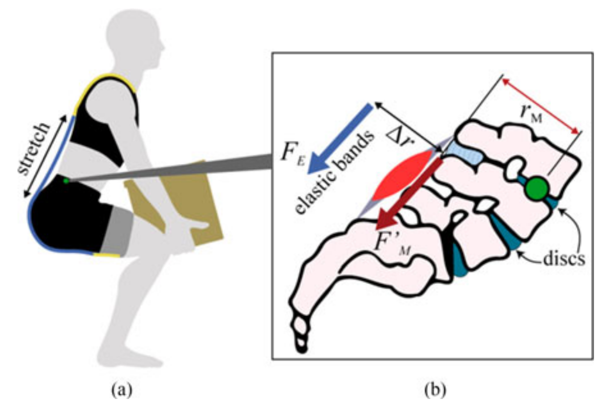

Fig. 3. (a) The biomechanically-assistive garment stretches during lifting and leaning. (b) Forces borne by the elastic bands are expected to offload the extensor muscles, and to reduce intervertebral disc loading by increasing the extensor moment arm (Δr). Analytically, from the moment balance equation, $F_E + F'_M < F_M$ from Fig. 1.

advantage over these tissues (Fig. 3). As a result, the elastic bands can provide equivalent extension moments at lower force magnitudes, thus reducing compression force on the lumbar spine [41].

B. Experimental Testing

Eight healthy subjects (7 male, 1 female, 74 ± 8.7 kg, 1.8 ± 0.05 m, 23 ± 3 yrs.) performed a series of leaning and lifting tasks with vs. without the prototype, while we recorded body kinematics, elastic band force and electromyography (EMG) data. The subjects, with no history of low back injury or pain in the past 6 months, provided informed, written consent prior to participating in this study. For each leaning task, subjects leaned forward for 30 seconds to pre-determined angles (30°, 60°, 90° from vertical) while holding a 4.5 kg weight (diameter: 20 cm, thickness: 3 cm) to their sternum. This weight approximated the additional flexion moment about the lumbar spine resulting from holding arms outstretched (e.g., a surgeon manipulating tools). A plumb line was used as a visual reference to assist subjects in reaching each pre-determined angle. For each lifting task, the subjects lifted a weight (12.7 kg and 24 kg) from the floor up to a standing posture (using squat lifting form) for 10 cycles, paced at 30 beats per minute (via metronome). Subjects performed lifting and leaning tasks without (control) and then while wearing the prototype. Before each task was performed with the prototype, the slack length of the elastic bands were manually adjusted such that peak elastic band forces were >20% of bodyweight during the task.

C. Data Collection

Body kinematics, EMG and elastic band forces were measured. Kinematics of the pelvis, trunk, and of the kettle-bell weight, were measured at 100 Hz with an optical motion capture system (Vicon). To track pelvic motion, retroreflective markers were placed bilaterally on the following landmarks: anterior superior iliac spine, iliac crest, posterior superior iliac spine, and greater trochanter. To track trunk motion, retroreflective markers were placed on the 7$^{\text{th}}$ cervical vertebrae, and bilaterally on the acromion and clavicle. The motion of the kettle-bell weight was tracked using a cluster of 4 markers on the handle. EMG of the bilateral lumbar erector spinae muscles (~30 mm bilateral to L3 spinal processes) were recorded at 2000 Hz using wireless sensors (Delsys). The erector spinae muscles were chosen because they generate the majority of the extensor moment during sagittal plane manual handling tasks [29], [40], and provide an estimate of general back muscle activity [8]. Tension forces in each elastic band were measured at 2000 Hz using two uniaxial load cells (Futek). All data were collected synchronously.

D. Data Processing

Motion data were low-pass filtered at 6 Hz, and load cell data at 15 Hz (each via 4th order dual-pass Butterworth filter). EMG signals were de-meaned, high-pass filtered at 30 Hz, rectified, and then low-pass filtered at 10 Hz, using 3rd order, dual-pass Butterworth filters [44]. EMG envelopes were then normalized to the maximum filtered EMG value observed across all trials (max observed activation).

E. Data Analysis

Mean normalized EMG was computed as the primary outcome metric for leaning and lifting tasks, and was used as an indicator of muscle loading. Mean EMG for leaning tasks was calculated by first averaging left and right lumbar erector spinae EMG, and then averaging the result over the duration of static leaning (Fig. 4(a)). A vector projected from the L5-S1 junction to the mid-point between the suprasternal notch and 7th cervical vertebrae was used to measure trunk angle. Subject-specific leaning trials were excluded from analysis if the maximum measured elastic band force was <20% of subject bodyweight (suggesting that the prototype may have shifted during the trial). Lifting tasks were divided into cycles. Each cycle was normalized to 1000 data points (due to differing cycle lengths), and then averaged. Mean EMG for lifting tasks was calculated by first averaging left and right erector spinae EMG, and then averaging the result over the lifting cycle duration (Fig. 4(a)). Peak EMG during the lifting cycle was also quantified, and was used as an indicator of peak loading, assuming a proportional relationship between EMG activity and muscle force. Average lifting cycle durations were calculated and compared for each subject across conditions with vs. without the prototype to ensure that cycle durations were consistent. Subject-specific lifting trials were excluded if the maximum elastic band force was <20% of subject bodyweight, if maximum or minimum trunk angles varied by >10°, or if displacement of the weight varied by >5 cm with vs. without the prototype. Intersubject means and standard deviations were calculated and paired t-tests were performed to evaluate differences in outcome metrics with vs. without the prototype ($\alpha = 0.05$).

F. Model-Based Estimates of Reduction in Disc Loading

To gain an insight into the effects of the biomechanically-assistive garment on intervertebral disc compression force, we

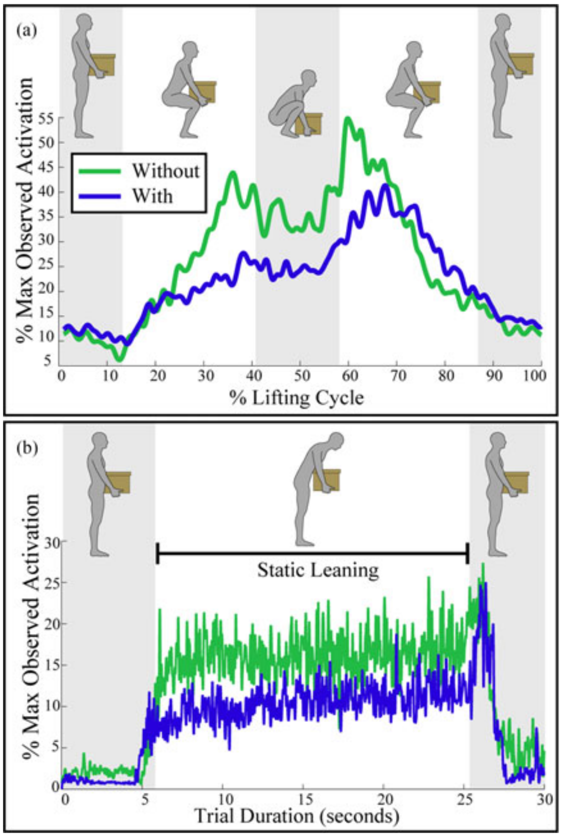

Fig. 4. Mean normalized EMG of the erector spinae muscles for a representative subject. (a) EMG for lifting task with vs. without the biomechanically-assistive garment prototype. Trials were parsed into cycles; a cycle began with the subject standing upright, next the subject squatted down, and then stood back upright to complete the cycle. (b) EMG for leaning task with vs. without the prototype. Mean EMG was calculated over the middle of the trial, during which the subject was leaning statically.

implemented a simple, analytical model of the lumbar spine during static leaning (similar to [39], [45]). The model assumes a static leaning posture, achieved with hip flexion and no lumbar flexion, at a fixed angle from the vertical ($\theta = 45°$). This results in a constant flexion moment about the L5-S1 intervertebral disc (~105 Nm) due to the weight of the HAT estimated using anthroprometrics of a 50th percentile male [46] (Fig. 1). For case I, the extension moment was assumed to be generated entirely by an extensor muscle force (F_M) acting with a 7 cm moment arm (r_M) [26]. Disc compression force (F_{disc}, acting axially along the spine) in case I was computed as the sum of the muscle force and the force due to the weight of the HAT ($F_{disc,I} = F_M + F_{HAT}\cos(\theta)$). In case II, we introduced another source of extension moment, due to forces in the elastic bands of the biomechanically-assistive garment (F_E, Fig. 3) acting at a larger moment arm than the muscles ($r_M + \Delta r$). We then updated the disc compression force calculation ($F_{disc,II} = F'_M + F_{HAT}\cos(\theta) + F_E$). Next, we varied Δr from 0 to 20 cm, and F_E from 0 to 380 N, and calculated the updated disc compression force. Reduction in disc compression, vs. case I (with no external assistance), was then computed as $(\frac{F_{disc,II} - F_{disc,I}}{F_{disc,I}} \cdot 100)$ and used to create a contour plot.

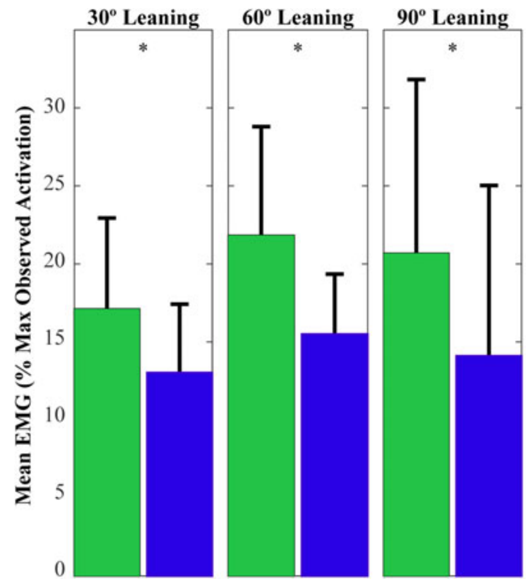

Fig. 5. Mean EMG of the erector spinae muscles was reduced during leaning when wearing the biomechanically-assistive garment prototype (dark blue) vs. not wearing it (green). Reductions were statistically significant for all leaning angles tested ($p < 0.05$, denoted with asterisks). Intersubject means and standard deviations are depicted.

TABLE I
SUBJECT-SPECIFIC RESPONSES

		Trials				
		30°	60°	90°	12.7 kg	24 kg
Subject	**1**	−26%	−18%	−88%	−11%	−16%
	2	−3%	−41%	−23%	N/A	N/A
	3	−35%	−14%	−18%	−15%	−14%
	4	−30%	−35%	−82%	N/A	N/A
	5	−9%	−29%	−26%	−13%	−22%
	6	−35%	−22%	−1%	−2%	−19%
	7	−27%	−33%	−33%	−30%	−22%
	8	N/A	N/A	−72%	−10%	−2%

Changes in subject-specific EMG, while wearing the biomechanically-assistive garment prototype, relative to not wearing it (control). N/A indicates subject-specific tasks that were excluded.

III. RESULTS

A. Leaning Task

When wearing the prototype, mean EMG was reduced by 23% ± 13% (p = 0.011), 27% ± 10% (p = 0.006) and 43% ± 33% (p = 0.001) for the 30°, 60° and 90° leaning tasks, respectively (Fig. 5). All subjects exhibited a reduction in EMG for all leaning angles, although the magnitude of reduction varied significantly from subject to subject (reduction range: 1-88%, Table I). Elastic band forces were 23% ± 3%, 26% ± 3%, and 30% ± 5% bodyweight for the 30°, 60° and 90° leaning tasks respectively. Mean leaning angles with vs. without the prototype were not significantly different for the 30° (p = 0.071), 60° (p = 0.847) or 90° (p = 0.121) tasks (Table II). One subject was excluded from the analysis of the 30° and 60° conditions because measured elastic band forces were <20% of their bodyweight.

TABLE II
TASK PERFORMANCE METRICS

	Lean 30°	Lean 60°	Lean 90°
	Trunk Angle	Trunk Angle	Trunk Angle
Without	28 ± 6°	54 ± 6°	91 ± 6°
With	31 ± 7°	55 ± 10°	89 ± 8°
		Lifting 12.7 kg	
	Trunk Angle	Cycle Time	Weight Height
Without	54 ± 11°	3.99 ± 0.01 s	54.2 ± 2.8 cm
With	53 ± 11°	3.99 ± 0.01 s	53.6 ± 2.5 cm
		Lifting 24 kg	
	Trunk Angle	Cycle Time	Weight Height
Without	50 ± 7°	3.99 ± 0.02 s	53.8 ± 2.9 cm
With	53 ± 11°	4.00 ± 0.01 s	53.0 ± 3.1 cm

Kinematic metrics and cycle time durations were comparable (i.e., not significantly different) when performing each task with vs. without the biomechanically-assistive garment.

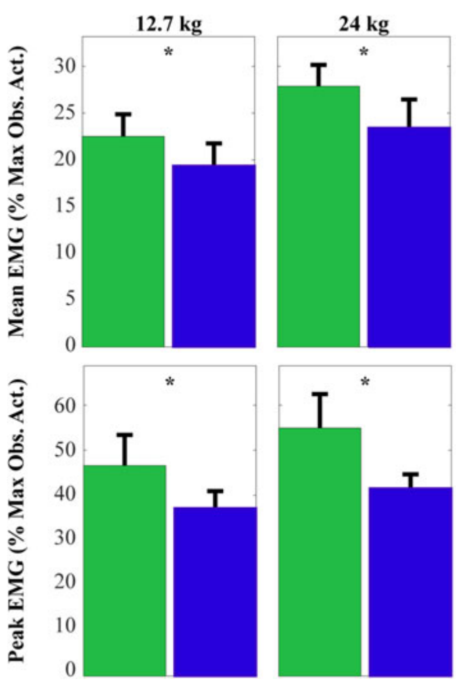

Fig. 6. Mean and peak EMG of the erector spinae muscles were reduced during lifting when wearing the biomechanically-assistive garment (dark blue) vs. when not wearing it (green). All reductions were statistically significant ($p < 0.05$, denoted with asterisks). Intersubject means and standard deviations are depicted.

B. Lifting Task

When wearing the prototype, mean EMG was reduced by 14% ± 9% ($p = 0.015$) and 16% ± 8% ($p = 0.004$) for the 12.7 kg and 24 kg lifting tasks, respectively. Peak EMG was reduced by 19% ± 13% ($p = 0.025$) and 23% ± 9% ($p = 0.004$) for the 12.7 kg and 24 kg lifting tasks, respectively (Fig. 6). All subjects exhibited EMG reductions while using the prototype to lift the kettle-bell weights (reduction range: 2-30%, Table I). Peak elastic band forces were 27% ± 4% and 27% ± 5% bodyweight for the 12.7 kg and 24 kg lifting tasks, respectively. On average across subjects, maximum trunk angles (12.7 kg: $p = 0.156$, 24 kg: $p = 0.224$), mean cycle duration times (12.7 kg: $p = 0.625$, 24 kg: $p = 0.413$) and kettle-bell displacements (12.7 kg: $p = 0.546$, 24 kg: $p = 0.111$) were not significantly different with vs. without the prototype (Table II). Two subjects were excluded from the analysis of the lifting tasks because there was >10° of difference in max/min trunk angles and >5 cm difference in kettle-bell displacement when comparing task performance with vs. without the prototype.

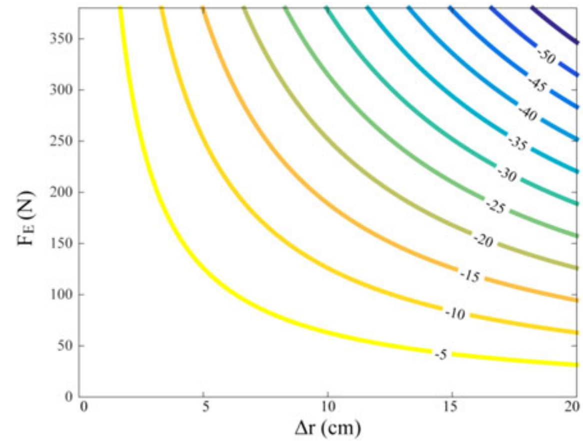

Fig. 7. Theoretical reduction in lumbar disc compression force when wearing a biomechanically-assistive garment. Disc force reduction increases as a function of increasing elastic cable force (F_E) and increasing moment arm (Δr).

C. Model-Based Estimates of Disc Loading

Model predictions, based on a simple moment balance, indicate that offloading the low back muscles with a biomechanically-assistive garment is expected to reduce intervertebral disc compression forces (Fig. 7). The magnitude of disc force reduction is expected to increase with both increasing elastic band force (F_E) and increasing moment arm (Δr).

IV. DISCUSSION

Here we introduce a new design concept for a biomechanically-assistive garment, and demonstrate its potential to mitigate forces experienced by the low back muscles and discs. We found that our biomechanically-assistive garment prototype offloaded the lumbar extensor muscles during both static leaning and dynamic lifting, as evidenced by reductions in EMG. The magnitudes of mean EMG reductions presented in this study (intersubject means ranging 14-43%) were comparable to reductions shown from previous exoskeletal devices with larger form-factors (~9-54%, [34], [35], [37], [41], [42]). Furthermore, the biomechanically-assistive garment demonstrated the ability to reduce both mean and peak EMG, which may affect risk of injuries due to repetitive (overuse) or high (overloading) forces. Model-based estimates indicate that the garment could also offload the intervertebral discs because the garment acts with a mechanical advantage over the extensor muscles. Overall, preliminary evidence strongly supports the feasibility of using low–profile, biomechanically-assistive garments to offload lumbar extensor muscles, which may help to reduce force-induced injury risks and/or fatigue during leaning and lifting tasks.

All subjects exhibited reduced lumbar erector spinae muscle activity when wearing the prototype; however, the magnitude of EMG reduction was subject and task-specific (Table I). Four of the eight subjects (subjects 1, 3, 4, and 7) exhibited reductions $\geq$10% for all leaning and lifting trials completed, suggesting that a subset of the population could derive broad and substantial benefits (in terms of offloading muscles) from a biomechanically-assistive garment. The four remaining subjects (subjects 2, 5, 6 and 8) exhibited reductions $\geq$10% for most tasks they completed, but $<$10% for other tasks. These results may reflect subject-specific adaptations (e.g., altered kinematics or muscle co-contractions), or variability in EMG recordings, or in how we pretensioned the prototype for each person and task. Overall these reductions suggest that the prototype consistently reduced loading of the lumbar extensor muscles.

The degree to which a biomechanically-assistive garment (and other similar wearable assistive devices) can reduce disc compression forces depends on the mechanical advantage (increased moment arm, Δr) it provides relative to the extensor muscles, and on the force the device applies to the user (F_E). There are practical limits for F_E and Δr: F_E is primarily limited by the amount of force that can be safely and comfortably applied to a person, whereas Δr is limited by the desired form-factor and tolerable size of the device; both of which may vary for a given individual, task and/or setting. Using the parameters of the prototype tested in this experiment ($F_E = \sim 200 - 300\ N$ and $\Delta r = \sim 1 - 4$ cm), reductions in disc loading of $\sim$5-10% would be expected. If a biomechanically-assistive garment were used on a daily basis or for extended periods of time, then relatively small reductions could provide cumulative offloading of the spine. Alternatively, modestly increasing the moment arm, via a permanent protruding structure (e.g., [39]) or an extensible mechanism, could serve to further offload the discs. In support of this, Abdoli-Eramaki *et al.* estimated that a similar device acting in parallel to the lumbar spine with $\Delta r = \sim$15 cm would reduce disc loading by $\sim$23-29% [39].

There were several limitations to this study. A relatively small sample (N=8) and subset of tasks were tested in this initial feasibility experiment; however, this was sufficient to demonstrate proof-of-concept for the biomechanically-assistive garment. This study only addressed short-term effects on the activity of isolated lumbar extensor muscles. It would be of value in future studies to measure additional back extensor muscles as well as quantify abdominal co-contraction (which can contribute to lumbar tissue loading [40], [47]), estimate compressive and shear loading on the lumbar spine (e.g., using optimization or EMG-assisted modeling techniques similar to previous work [27], [40], [48], [49]), and to explore longer-term effects. In this work, muscle activity was normalized to the maximum observed activation across all trials. Relative changes in EMG were then interpreted to assess the feasibility of the biomechanically-assistive garments to offload the low back; however, there is limited ability to draw conclusions about clinical significance at this time. Further studies are needed. If future testing is performed on a population with a history, or elevated risk, of back pain, then it would also be beneficial to collect self-reported pain metrics or to track injuries over time; though the latter would require a larger longitudinal study. The moment balance model used to estimate relative changes in disc compression forces assumed anatomical simplicity and did not explicitly model muscle co-contraction, passive tissues or other segmental dynamics. Nevertheless, this model provides an order-of-magnitude approximation of expected force reduction and insight regarding the sensitivity of disc compression force to device design parameters. Though these approximations would benefit from further or more direct (invasive) validation, conclusions in this study do not depend on model force predictions being highly accurate. The prototype tested in this study required us to manually adjust elastic band tension for each person and task. Future prototypes may integrate quasi-passive clutching capabilities (to engage/disengage the elastic elements) and wearable sensors in order to more precisely control the magnitude and timing of assistance. Alternatively, further assistance and versatility might be provided using a portable actuator unit, similar to robotic exosuit hardware that has been developed to assist walking [50], [51]. Finally, we note that for all physical interventions, the act of redistributing forces (in this case, applying forces to the shoulders and thighs to offload the spine) may have unexpected or unintended consequences in terms of altering loading elsewhere in the body. These concomitant biomechanical effects warrant investigation in future studies.

Biomechanically-assistive garments may have broad societal appeal as a lightweight, unobtrusive and cost-effective means of reducing low back loading, which could affect the risk of low back injury and pain in daily life. The ability to non-invasively alter lumbar loading and muscular effort may also have applications for reducing fatigue (which is distinct from, but can be associated with discomfort or low back pain [23]) or aiding in patient recovery post-injury. These garments provide a unique means to mitigate low back loading by adapting clothing, which is ubiquitous and already worn every day. These garments are expected to be cost-effective because they can leverage existing soft goods manufacturing methods and infrastructure. Moving forward, biomechanically-assistive garments could also be instrumented with wearable sensors and smart algorithms to control quasi-passive structures (e.g., elastic bands that can be clutched or unclutched to selectively provide assistance during certain tasks). One future possibility would be for this instrumented clothing to also provide users with actionable biofeedback about their low back loading (similar to how pedometers track and report steps per day), and to alert users to high loading or predicted injury risks.

V. Conclusion

We found that a passive, biomechanically-assistive garment was able to offload low back muscles during leaning and lifting. These findings support the feasibility of using biomechanically-assistive garments to reduce low back injury and resultant pain due to high (overloading) and/or repetitive (overuse) forces on lumbar musculature**.**

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