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Design and pilot evaluation of a reconfigurable spinal exoskeleton

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Abstract

Low back pain is a leading cause of disability, and there is a tremendous need for nonsurgical, nonpharmaceutical interventions to manage it. Versatile spinal exoskeletons have been proposed as a method of supporting or augmenting the wearer, but experimental data from human subjects are limited, and the effects of such exoskeletons remain poorly understood. We thus present a prototype of a reconfigurable spinal exoskeleton that features easily adjustable resistance and compression at multiple spinal levels, allowing us to study the effect of different exoskeleton configurations on the body. In a pilot evaluation with a single subject, both thoracic and abdominal compression were found to affect trunk angle, low back moment and the electromyogram of the erector spinae, though different exoskeleton configurations had different effects during different tasks. This supports the premise that intelligent mechanical adjustments of a spinal exoskeleton are necessary for optimal support or augmentation of the wearer, though the results need to be examined in a larger, varied sample of subjects.

I. Introduction

Low back pain (LBP) is a leading cause of disability worldwide [1], with lifetime prevalence estimates as high as 84% [2]. Treatment of LBP, however, is difficult, as it is a complex process with anatomical, physiological, neuromuscular and psychosocial components [3], [4]. Furthermore, most patients who visit doctors' offices with complaints of LBP are not found to have discernable tissue damage or injury with the use of current diagnostic techniques, and are diagnosed as patients with nonspecific LBP [2], [5]. With no clear

pathoanatomical cause, management of nonspecific LBP focuses on reducing pain and its consequences [5].

Methods for LBP management include surgery, analgesics, external immobilization/support (bracing), activity modification, and muscle strengthening. However, surgical and pharmaceutical methods should only be used in serious cases where other approaches prove ineffective, as there is great potential for side effects [6], [7]. There is thus an enormous need for nonsurgical and nonpharmaceutical interventions that could effectively manage LBP.

One promising LBP intervention is an orthosis (external brace) that aims to support the spine. However, while targeted motion limitation, load reduction and enhanced muscle activation are acknowledged to be important factors for LBP intervention, the ability of traditional rigid spinal orthoses to achieve these functions is highly contested [7]. For example, an orthosis designed to limit spinal motion may reduce this motion for some spinal segments but increase it for others; furthermore, while an orthosis may reduce spinal motion during some activities, it may not reduce it for others [8]-[10].

To address the weaknesses of traditional spinal orthoses, researchers have proposed, but have yet to fully realize, versatile spinal exoskeletons that can perform mechanical adjustments at multiple spine levels and in multiple movement directions [11]—[15]. Such exoskeletons could be passive or could be augmented with motors and sensors, potentially providing activity-specific spinal support in the same way that limb exoskeletons intelligently tailor their assistance to the wearer's activities [16]. Such flexible, versatile spinal exoskeletons therefore have the potential to better support the spine and consequently better manage LBP than traditional rigid spinal orthoses.

To support the development of such spinal exoskeletons, however, it is critical to know how different mechanical aspects such as compression and range of motion at different spinal levels affect the human body during different activities. Evaluations of traditional rigid spinal orthoses have shown that different designs have different effects on the body [9], [10], but it is unclear what mechanical aspects contribute to these differences and how. Some authors have applied numerical models to estimate these effects [17], but experimental data from human subjects remain limited.

Inspired by research into modular limb exoskeletons [18], our research consortium has developed a prototype articulated spinal exoskeleton whose mechanical aspects (compression, resistance, range of motion) can be adjusted quickly and easily. Our long-term goal is to augment the device with actuators and use it for therapeutic purposes. However, we will first use the device to conduct systematic investigations about how different mechanical aspects of spinal exoskeletons affect the human body during different activities, thus providing valuable information about how spinal exoskeletons can best support and/or augment the wearer. In this paper, we present the mechanical design of our spinal exoskeleton as well as the first, preliminary test with one human subject without LBP.

II. Materials and Methods

A. Spinal exoskeleton design

The spinal exoskeleton used in our research (Fig. 1) is a variable-resistance exoskeleton-Thoracic-Lumbar-Sacral orthosis that weighs 5 pounds. Its main section is a reconfigurable articulated exoskeletal spinal column (Fig. 1, center) that was designed to be easily adjustable and facilitate selected variable resistance (damping and friction) along the posterior of the torso. The main components of the column incorporate variable-segment axial (sagittal, transverse, longitudinal) resistance couplings as well as variable angular couplings. Curvic couplings integrated within the coupling assemblies allow selection of the angle between adjacent segments. We developed a viscoelastic coupling [19] to allow independent control of each joint's resistance and range of motion while avoiding the vibrational effects typically exerted along the spine by standard elastic springs. In addition, the lengths of the different segments can be easily adjusted to accommodate the individual wearer.

The locations of the viscoelastic couplings in the spinal column correspond to the L5-S1, L2-L3, T12-L1, and T7-T8 borders. A schematic of the device's couplings and segments as well as a detailed picture of a coupling assembly are shown in Fig. 2. The couplings and connecting segments are made of a combination of 3D-printed metal and nylon components, and are surrounded by a flexible housing. This housing is a combination of 90A urethane, viscoelastic material (sorbothane) and foam (similar to that used in shoe insoles). This flexible material is used at points where the device is attached to the body. Furthermore, rigid segments of the device conform to bony landmarks, and padding is used to distribute the orthosis load. The design avoids areas where point loading could lead to puncture.

Multiple trunk-grasping end-effectors are attached to the spinal column (Fig. 1, center) and connect to front parts of the exoskeleton using detachable straps (Fig. 1, right). The end-effectors are padded with silicone and foam for safety and comfort, and easily interchangeable abdominal sections of different sizes were developed to accommodate large variations in stomach size. Furthermore, the adjustable straps provide independently adjustable compression in the thoracic and abdominal regions. Investigating the effects of this thoracic and abdominal compression was the main goal of our pilot evaluation.

B. Pilot Evaluation Protocol

We evaluated two key biomechanical effects of the exoskeleton: 1) the increase or decrease in the demand of the back muscles and 2) how changing the thoracic and abdominal compression of the exoskeleton affects the trunk kinematics and low back kinetics of the wearer. For this purpose, a 32-year-old male subject (height 186 cm, weight 90 kg) who had not previously worn a spinal exoskeleton and had not experienced chronic LBP was recruited to participate in a study involving common everyday tasks. Photos of the subject wearing the exoskeleton are shown in Fig. 3. The experimental procedure was approved by the University of Wyoming Institutional Review Board.

The subject performed the following everyday tasks: walking in a straight line on level ground, walking up two steps, lifting a small box (20 pounds, 37×26×16 cm) from the

ground to abdominal height in a standing position, and standing up from a chair. These tasks were performed in three different exoskeleton configurations: high thoracic and high abdominal compression, high thoracic and low abdominal compression, and low thoracic and high abdominal compression. Furthermore, they were also performed without the exoskeleton.

To evaluate muscle demand, the electromyogram (EMG) of the left and right erector spinae was collected with a sampling frequency of 2 kHz using the Delsys Myomonitor wireless EMG system (Delsys Inc., MA). Bipolar electrodes were placed at L3 height, approximately 4 cm left and right from the midline of the spine, and a harness was placed above the subject's pelvis to hold the EMG transmitter. A similar setup was used in a previous evaluation of spinal exoskeletons [14], though that paper also measured muscles other than the erector spinae. To evaluate trunk kinematics, retroreflective markers were attached to the subject's hips, legs, shoulders, arms and head, then tracked with a sampling frequency of 160 Hz using six Vicon Bonita optical cameras (Vicon Motion Systems, UK).

The study protocol was as follows: The exoskeleton and sensors were placed on the subject, and the exoskeleton was set to the first configuration (high thoracic, high abdominal compression). The subject performed all four tasks in a random order, then performed all four tasks two more times in different random orders. The exoskeleton was then set to the second configuration, and the subject again performed all four tasks three times in random orders. The tasks were then performed in the third exoskeleton configuration using the same protocol, followed by the control condition of not wearing the exoskeleton. Finally, EMG was measured during maximum voluntary contraction conditions by having the subject lie on his stomach and try to lift his upper body (without using his arms) while an experimenter pushed down on the subject's shoulders with a steady force.

C. Data analysis

A three-dimensional linked segment model was constructed from marker data using the method of Kingma et al. [20], which also allowed moments to be estimated using a top-down approach. The trunk flexion angle and the low back extension moment were extracted from the segment model, and the peak low back moment and peak trunk angle (from the initial standing position) were calculated for each exoskeleton configuration and each task.

EMG was segmented by identifying the beginning and end of a task (e.g. box lift) from kinematic data. Segmented EMG signals were then detrended and filtered with a second-order 20–450 Hz bandpass filter. The signals were then rectified, and the envelope was extracted by applying a second-order lowpass filter with a cutoff frequency of 10 Hz. Each EMG envelope was normalized by the maximum value observed during maximum voluntary contraction conditions. The peak value of the EMG envelope was calculated for each exoskeleton configuration and each task.

III. Results

Peak low back moment and peak trunk angle are shown in Tables I and II for each exoskeleton configuration and each task. Furthermore, Figs. 4 and 5 show examples of EMG

signals from representative trials of lifting a box (Fig. 4) and standing up from a chair (Fig. 5).

IV. Discussion

Our preliminary results indicate that the reconfigurable exoskeleton modifies the kinematics and EMG of the human body: for example, during box lifting, one exoskeleton configuration reduced peak EMG of the erector spinae by about 30% compared to not wearing the exoskeleton (Fig. 4). The different configurations also have different effects. For example, Table I shows that the configuration with high thoracic and high abdominal compression reduced peak low back extension moment while the configuration with high thoracic and low abdominal compression increased peak low back extension moment during box lifting compared to not wearing the exoskeleton.

Furthermore, the three exoskeleton configurations had different biomechanical effects across the tasks: for example, while the configuration with high thoracic and low abdominal compression decreased peak trunk angles during walking on level ground and sit-to-stand motions, this decrease was not observed during the other two tasks (Table II). Similarly, while the configuration with low thoracic and high abdominal compression decreased peak EMG during box lifting (Fig. 4), it actually increased peak EMG during sit-to-stand motions (Fig. 5). This supports the premise that a traditional rigid spinal orthosis does not have beneficial effects for all activities, and that an adjustable spinal exoskeleton could provide tailored support for different activities.

Because these data were obtained from a single subject in a single session, we cannot generalize the findings or comment on repeatability. The effects of a spinal exoskeleton likely depend on the wearer's build, and we will continue the study by recruiting a larger number of subjects of different ages, biological sexes, proportions etc. Furthermore, we would also like to study the effect of the exoskeleton on subjects with chronic LBP. While most people with chronic LBP have no discernable tissue damage [2], [5], it would be very useful to examine the effect of the exoskeleton on people with, for example, degenerative disc disease.

Our future research will also expand and improve the study protocol. For example, we only studied the exoskeleton's effect on a single muscle (the erector spinae). This is not optimal, as reducing the demand on one muscle may shift demand to other muscles through regional interdependence [21], and we should thus also measure the effects of different configurations on, e.g., the abdominal muscles (as done by, e.g., Huysamen et al. [14]). It would also be useful to include other tasks in the study protocol, such as maintaining postural stability in response to unexpected perturbations. Finally, our pilot evaluation only examined the effects of thoracic and abdominal compression, but many other mechanical aspects of the exoskeleton remain to be investigated. This primarily includes the resistance of the different couplings, which can be easily adjusted and may have a strong effect on the body (as suggested by, e.g., Utter et al. [10]).

V. Conclusion

We presented a prototype of a reconfigurable spinal exoskeleton that features easily adjustable resistance, range of motion and compression at different spinal levels, allowing the study of the effects of different exoskeleton configurations on the human body. A preliminary evaluation performed with a single human subject demonstrated that adjusting thoracic and abdominal compression affects the electromyogram of the erector spinae as well as low back moments and trunk angles. Notably, no exoskeleton configuration has the same effect during observed tasks, suggesting that intelligently adjusting a spinal exoskeleton's configuration for different tasks (using, e.g., sensors and actuators as suggested by Huysamen et al. [14]) would be able to provide more effective support or augmentation for the wearer. In the future, the device will be used to determine how different mechanical aspects of spinal exoskeletons affect wearers with different builds, thus providing valuable information about how spinal exoskeletons can best support or augment the wearer.

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Figure 1.

The spinal exoskeleton used in our research. Left: full device. Center: spinal column with trunk-grasping end-effectors. Right: thoracic and abdominal front sections, which attach to the end-effectors using straps.

Figure 2.

Left: schematic of the spinal column's segments and viscoelastic couplings. Right: close-up picture of a coupling assembly.

Figure 3.
Rear and front views of a subject wearing the spinal exoskeleton.

Figure 4.

Electromyogram of the right erector spinae when lifting a box, for each exoskeleton configuration. At $t = 0$, the subject begins lifting the box off the floor.

Figure 5.

Electromyogram of the right erector spinae when standing up from a chair, for each exoskeleton configuration

TABLE I.

PEAK LOW BACK EXTENSION MOMENT FOR EACH EXOSKELETON CONFIGURATION AND EACH TASK. ALL VALUES ARE GIVEN IN NM. TC = THORACIC COMPRESSION, AC = ABDOMINAL COMPRESSION.

Configuration	Task			
	<i>Walk</i>	<i>Steps</i>	<i>Box lift</i>	<i>Sit-to-stand</i>
No exoskeleton	20.5	39.2	142.2	195.0
High TC, high AC	25.3	41.7	133.1	186.6
High TC, low AC	18.2	41.1	147.1	188.0
Low TC, high AC	25.5	38.3	137.4	205.7

TABLE II.

PEAK TRUNK FLEXION ANGLES FOR EACH EXOSKELETON CONFIGURATION AND EACH TASK. ALL VALUES ARE GIVEN IN DEGREES. TC = THORACIC COMPRESSION, AC = ABDOMINAL COMPRESSION.

Configuration	Task			
	<i>Walk</i>	<i>Steps</i>	<i>Box lift</i>	<i>Sit-to-stand</i>
No exoskeleton	12.1	17.2	30.4	52.5
High TC, high AC	7.8	17.4	38.9	59.0
High TC, low AC	9.8	17.6	33.0	40.8
Low TC, high AC	12.7	15.9	30.2	48.5