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Simultaneously varying back stiffness and trunk compression in a passive trunk exoskeleton during different activities: A pilot study

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Abstract— **Passive trunk exoskeletons support the human body with mechanical elements like springs and trunk compression, allowing them to guide motion and relieve the load on the spine. However, to provide appropriate support, elements of the exoskeleton (e.g., degree of compression) should be intelligently adapted to the current task. As it is not currently clear how adjusting different exoskeleton elements affects the wearer, this study preliminarily examines the effects of simultaneously adjusting both exoskeletal spinal column stiffness and trunk compression in a passive trunk exoskeleton. Six participants performed four dynamic tasks (walking, sit-to-stand, lifting a 20-lb box, lifting a 40-lb box) and experienced unexpected perturbations both without the exoskeleton and in six exoskeleton configurations corresponding to two compression levels and three stiffness levels. While results are preliminary due to the small sample size and relatively small increases in stiffness, they indicate that both compression and stiffness may affect kinematics and electromyography, that the effects may differ between activities, and that there may be interaction effects between stiffness and compression. As the next step, we will conduct a larger study with the same protocol more participants and larger stiffness increases to systematically evaluate the effects of different exoskeleton characteristics on the wearer.**

Clinical Relevance— **Trunk exoskeletons can support wearers during a variety of different tasks, but their configuration may need to be intelligently adjusted to provide appropriate support. This pilot study provides information about the effects of exoskeleton back stiffness and trunk compression on the wearer, which can be used as a basis for more effective device design and usage.**

I. Introduction

Trunk exoskeletons are an emerging type of technology with great promise for human augmentation and rehabilitation. Such exoskeletons are designed to reduce the load on the spine and guide trunk motion, thus reducing back pain and improving stability for people with back injuries [1] as well as reducing the risk of back injury for workers in physically demanding occupations [2]. Since back pain represents a major cause of disability worldwide [3], trunk exoskeletons could thus significantly improve human health.

Existing trunk exoskeletons can be roughly divided into active devices, which use motors to apply torques to the limbs and augment the wearer's movements [4], [5], and passive devices, which have no motors and instead rely on elements such as elastic springs/bands (to store and release energy) and trunk compression (to restrict undesirable movements) [6]–[8]. While passive devices are lighter, simpler and cheaper than active devices, they must also be carefully designed and adjusted for the user's body and specific activity, as studies have found that some passive exoskeleton configurations are more appropriate for some activities than others [7]–[9]. Similar results have previously been observed in passive rigid spinal orthoses (braces) [10], [11].

To provide an adjustable and personalized experience for wearers, several research groups have developed trunk exoskeletons with manually adjustable elements (e.g., trunk compression [7], spring pretension [12]) or multiple "modes" that the user can switch between [9]. Recently, both our research group [7] and other groups [6], [13] have introduced the concept of "semi-active" exoskeletons where the adjustment or mode-switching could instead be done automatically by onboard sensors and micromotors that would detect the user's current activity and adjust the exoskeleton appropriately. This could potentially provide optimal assistance and allow the user to cognitively focus on the task rather than on exoskeleton adjustment. However, to effectively perform either manual or automated adjustment, it is necessary to know how different exoskeleton characteristics affect the wearer.

In our previous study, we introduced a prototype adjustable trunk exoskeleton and systematically studied the effect of varying trunk compression at thoracic and abdominal levels during multiple activities [7]. We found that both thoracic and abdominal compression affect kinematics and trunk muscle electromyograms (EMG), with thoracic compression having different effects than abdominal compression. Furthermore, different effects were observed in different activities, indicating that some compression settings are more suitable for specific activities than others. This supports the premise of intelligently adjusting a trunk exoskeleton to specific activities to achieve optimal support for the wearer. However, trunk compression is not the only exoskeleton characteristic that may be relevant. Furthermore, individual characteristics should not be studied in isolation, as there may be important interplay between different characteristics.

In this paper, we present a pilot study on simultaneously varying both trunk compression (as done in the previous study [7]) and the stiffness of the exoskeletal spinal column in our adjustable trunk exoskeleton. While we plan to conduct a study with a larger change in stiffness in the future, this pilot study was done with a small change in stiffness to verify the protocol and ensure that a small increase is not unpleasant or unsafe for participants.

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II. Materials and Methods

A. Hardware and Software

The passive trunk exoskeleton used as the basis for this work is shown in Fig. 1. It was originally developed by Livity Technologies (Highlands Ranch, USA), weighs 3.7 kg, and consists of multiple sections: an exoskeletal spinal column (with seven variable couplings at four trunk levels), thoracic and abdominal front modules, six trunk-grasping end-effectors that run from the spinal column to the front modules (two over the shoulder, two at the vertical midpoint of the trunk, and two at the hips), and elastic straps that connect the front modules to the end-effectors. More details about these aspects are available in our previous study [7].

Trunk compression can be varied independently at thoracic and abdominal levels by manually changing the tightness of the elastic straps that connect the front modules to the trunk-grasping end-effectors. The straps are ladder straps threaded through ratchet buckles, and thus have multiple discrete tightness levels. While compression can be varied independently at thoracic and abdominal levels, in this study we only varied it at both levels simultaneously (i.e., never changed it at only one level), as independent adjustments were already examined in our previous study [7].

Exoskeletal spinal column stiffness can be varied by changing the stiffness of the seven couplings located at four trunk levels (one at T7-T8, three at T12-L1, one at L2-L3, two at L5-S1 – Fig. 2). Each coupling consists of fixed and variable stiffness subassemblies in series, which have both elasticity and viscosity. The rotor is connected to the stator via folded leaf-springs modeled as rigid links with revolute joints and torsional springs located such that they accurately describe the motion of the rotor. Additionally, the rotor is linked to the stator using small radially extending volute spring "arms" that can be engaged and disengaged (Fig. 3). In our previous study [7], these couplings were kept constant at a moment of resistance of 4.7 N·m, a torsional stiffness of 180 N·m/rad, and a viscosity of 1350 Pa·s. In the current study, we varied the stiffness of four couplings – two at the T12-L1 level and two at the L5-S1 level, responsible for rotation in the sagittal and frontal planes at that level. The stiffness of each coupling was manually switched between "not stiff" (without the volute springs engaged, same characteristics as above) and "stiff" (with volute springs engaged, ~10% stiffness increase).

B. Participants

Six individuals (1 woman, 5 men) with no history of chronic low back pain or back injury took part in the study. They were 22-29 years old (median 24), with heights of 173-189 cm (median 179 cm) and weights of 66-91 kg (median 78 kg). All self-reported that they were right-handed, and all signed an informed consent form after having the purpose and procedure of the study explained to them.

C. Study Protocol

The pilot evaluation protocol was approved by the University of Wyoming Institutional Review Board (protocol #20200129DN02643) and was an expanded version of the protocol from our previous study [7]. Each participant attended two sessions in the Biomechanics Laboratory of the University of Wyoming on two separate days.

Possible exoskeleton configurations: In each session, the participant wore the trunk exoskeleton in six configurations corresponding to all possible combinations of two compression settings (both thoracic and abdominal compression high vs. both thoracic and abdominal compression low) and three stiffness settings (all four couplings stiff, lower two couplings stiff, all four couplings not stiff). Both high and low compression were set differently for each participant based entirely on that participant's subjective perception: high compression was the tightest elastic strap setting that was still considered comfortable by the participant while low compression was the loosest setting that was still perceived as pressure by the participant. Stiffness was varied between "stiff" and "not stiff" by engaging or disengaging the volute springs. Furthermore, there was also a "no exoskeleton" condition in both sessions.

Figure 1. The trunk exoskeleton worn by a person, front and back.

Figure 2. Close-ups of the couplings at different spinal column levels.

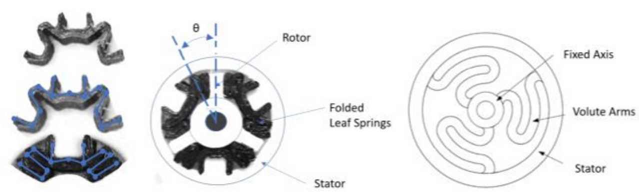

Figure 3. Left: the loaded and unloaded folded leaf spring chains. Middle and right: the fixed (middle) and variable stiffness (right) subassemblies.

Order of conditions within session: In both sessions, the "no exoskeleton" condition was always first or last, and the three stiffness settings were then applied in order from "all stiff" to "all not stiff" or vice-versa, with both compression settings tested in random order within each stiffness condition. This was done to minimize the amount of time spent varying stiffness and removing the exoskeleton, which are more time-consuming than varying compression.

Session 1: Participants were asked to complete multiple activities both without the exoskeleton and with the exoskeleton in all six compression/stiffness configurations. The activities were:

- Walking across the lab in a straight line at a self-selected moderate pace.
- Standing up from an initial sitting position on a padded stool, with arms crossed on the chest.
- Lifting a 20-pound (9.1 kg) box from the floor in front of the participant with both arms to waist level. No specific lifting strategy was prescribed, and the box's initial position was approximately 30 cm in front of the participant's feet.
- Lifting a 40-pound (18.2 kg) box from the floor in front of the participant with both arms to waist level.

Session 2: Participants sat on a padded stool with their eyes closed while perturbations were applied to their trunk, testing their ability to maintain stability. To apply perturbations, a rope was tied around the trunk at armpit height, led over a metal bar, and attached to a 20-pound (9.1 kg) weight that hung in the air. For each repetition, the participant first sat with eyes closed for 5-10 seconds while the weight hung in the air; at a randomly chosen moment, the rope was then disconnected from the weight using a mechanical release, and the participant had to compensate for this unexpected perturbation. In each exoskeleton configuration, perturbations were applied three times: with the rope and weight attached from the front, back, or right (dominant) side. This procedure was adapted from a previous biomechanics study on predicting knee injury [14] and also used in our previous work [7].

At the end of each session, maximum voluntary contraction (MVC) tests were performed to obtain maximum EMG values for all measured muscles (see next section). This was done by having the participant first lie on their stomach and try to lift their upper body for 5 s while the experimenter pushed down on their shoulders, then having the participant lie on their back and try to lift their upper body for 5 s while the experimenter pushed down on their shoulders. The same procedure was used in our previous work [7].

D. Measurements and Data Analysis

Two measurement types were taken: trunk kinematics and EMG. Kinematics were measured using eight Vicon Bonita cameras (Vicon Motion Systems, UK) and retroreflective markers at 160 Hz. In the first session, the markers were placed on the vertex, gonions, acromioclavicular joints, olecranon processes, midpoints of radial and ulnar styloid processes, third metacarpal heads, anterior superior iliac spines, posterior superior iliac spines, iliac crests and heels; additionally, a marker was placed on the estimated center of mass of the exoskeleton and two markers were placed on the box used in the lifting activity. In the second session, two markers were placed on the anterior superior iliac spines and two were placed on the acromioclavicular joints; additionally, one marker was attached to the weight to detect the exact time of perturbation. EMG was collected using the Trigno Avanti wireless system (Delsys Inc, Boston, MA) at 1926 Hz from four muscles: the left and right erector spinae (ES) and the left and right rectus abdominis (RA).

Segmentation: In both sessions, signals were manually segmented into individual trials – activity repetitions in the first session and individual perturbations in the second session. In the first session, a walking trial was segmented based on two heel-ground contact events while other activities were segmented based on the maximal and minimal values of the middle point of the two iliac crests. In the second session, a trial began when the weight was dropped and ended when participants stopped moving.

EMG analysis in both sessions began by detrending the segmented signals, filtering with a fourth-order 20-450 Hz bandpass filter, and rectification. A fourth-order 10 Hz lowpass filter was then used to obtain the EMG envelope, which was normalized by dividing it by that participant's MVC value for that muscle and that session. Finally, mean and peak values of each envelope were used as outcome variables.

Kinematic analysis in session 1: Marker data were filtered with a low-pass filter at 15 Hz, and a 3-dimensional linked segment model was constructed from marker data using the method of Kingma et al. [15]. Low back flexion angles were calculated between the upper trunk reference frame and the pelvis reference frame, and low back extension moments were calculated using a top-down inverse dynamic model [15]. For each trial, mean and peak flexion angles as well as mean and peak extension moments were then used as outcome variables.

Kinematic analysis in session 2: The segment model was constructed as in session 1. The trunk vector was then defined by the middle point of the two shoulders and the middle point of the two iliac crests. The trunk angle was calculated as the angle of the trunk vector relative to the trunk vector at the start of the trial. Finally, peak trunk deflection (maximum angle difference from starting orientation) and trunk deflection 150 ms after the start of the trial were used as outcome variables.

Result presentation: Results are given as means ± standard deviations. Due to the preliminary nature of the study, no statistical tests were performed, though we acknowledge that results are somewhat unreliable given the small sample size.

III. RESULTS

Table I shows results of the first session for four outcome variables in all dynamic tasks and all exoskeleton configurations. Table II then shows results of the second session for four outcome variables in all perturbation directions and all exoskeleton configurations.

While the results are not statistically significant given the small sample size, some differences between exoskeleton configurations can be seen during both dynamic tasks (session 1) and when reacting to unexpected perturbations (session 2). Furthermore, there do appear to be interaction effects between compression and stiffness, and effects of the exoskeleton do appear to vary from task to task. Examples of changes in EMG and kinematics as a result of changes in exoskeleton configuration are highlighted and discussed in the next section.

TABLE I. MEANS ± STANDARD DEVIATIONS FOR FOUR OUTCOME VARIABLES DURING FOUR DYNAMIC TASKS: WALKING (WALK), SIT-TO-STAND (S2S), LIFTING 20 LB (20LB), AND LIFTING 40 LB (40LB). COLUMNS REPRESENT DIFFERENT EXOSKELETON CONFIGURATIONS. NO EXO = NO EXOSKELETON. COMPRESSION SETTINGS: LC = LOW COMPRESSION, HC = HIGH COMPRESSION. STIFFNESS SETTINGS: NS = NOT STIFF, LS = LOWER TWO COUPLINGS STIFF, AS = ALL FOUR COUPLINGS STIFF. EMG = ELECTROMYOGRAM, RA = RECTUS ABDOMINIS, ES = ERECTOR SPINAE, MVC = MAXIMUM VOLUNTARY CONTRACTION.

		No Exo	LC-NS	LC-LS	LC-AS	HC-NS	HC-LS	HC-AS
peak flexion angle (degrees)	walk	4.6 ± 5.5	8.2 ± 5.9	4.7 ± 6.0	6.5 ± 7.4	4.9 ± 5.3	7.3 ± 3.7	7.6 ± 5.1
	S2S	10.5 ± 5.0	12.8 ± 7.8	14.7 ± 7.5	12.8 ± 8.4	14.5 ± 5.6	14.1 ± 6.0	17.3 ± 7.3
	20lb	24.0 ± 6.0	32.7 ± 3.0	31.9 ± 6.0	30.1 ± 6.9	30.6 ± 3.2	30.9 ± 2.0	29.9 ± 4.5
	40lb	23.0 ± 4.5	30.4 ± 4.2	30.4 ± 7.3	30.0 ± 6.0	31.8 ± 4.2	28.9 ± 3.4	28.6 ± 4.6
peak extension moment (Nm)	walk	-28 ± 12	-19 ± 12	-20 ± 10	-18 ± 13	-18 ± 13	-24 ± 19	-22 ± 20
	S2S	-98 ± 234	-98 ± 17	-102 ± 25	-94 ± 22	-103 ± 17	-100 ± 23	-108 ± 28
	20lb	-195 ± 26	-180 ± 23	-176 ± 20	-176 ± 20	-179 ± 24	-184 ± 37	-178 ± 31
	40lb	-240 ± 24	-220 ± 23	-220 ± 16	-227 ± 18	-222 ± 28	-231 ± 30	-227 ± 32
peak right RA EMG (% MVC)	walk	4.4 ± 2.9	4.2 ± 2.3	4.2 ± 1.7	4.4 ± 1.3	3.7 ± 2.0	3.9 ± 2.1	4.8 ± 3.1
	S2S	2.8 ± 1.6	2.6 ± 1.6	3.0 ± 2.2	3.1 ± 2.3	2.7 ± 1.6	3.8 ± 2.8	3.8 ± 2.6
	20lb	4.0 ± 3.2	3.9 ± 2.6	3.6 ± 2.3	4.0 ± 3.4	3.5 ± 2.4	3.9 ± 2.4	3.9 ± 3.2
	40lb	4.2 ± 4.1	3.7 ± 2.2	3.8 ± 2.4	4.4 ± 3.2	4.3 ± 2.8	4.3 ± 2.4	5.7 ± 4.8
peak left ES EMG (% MVC)	walk	16.2 ± 10.3	12.0 ± 5.2	12.8 ± 7.4	9.2 ± 4.7	14.8 ± 6.3	14.7 ± 6.9	15.1 ± 8.7
	S2S	23.3 ± 5.8	21.6 ± 4.3	24.6 ± 4.7	20.1 ± 4.5	24.8 ± 8.5	28.8 ± 8.9	32.8 ± 9.8
	20lb	62.2 ± 24.8	60.5 ± 31.4	54.9 ± 22.2	54.2 ± 27.5	51.4 ± 21.0	50.6 ± 13.8	58.1 ± 21.1
	40lb	66.5 ± 15.0	64.7 ± 18.9	79.6 ± 22.2	67.6 ± 22.9	66.9 ± 27.4	88.3 ± 29.8	78.4 ± 25.9

TABLE II. MEANS ± STANDARD DEVIATIONS FOR FOUR OUTCOME VARIABLES DURING PERTURBATIONS FROM THE FRONT (F), BACK (B), AND SIDE (S). COLUMNS REPRESENT DIFFERENT EXOSKELETON CONFIGURATIONS. NO EXO = NO EXOSKELETON. COMPRESSION SETTINGS: LC = LOW COMPRESSION, HC = HIGH COMPRESSION. STIFFNESS SETTINGS: NS = NOT STIFF, LS = LOWER TWO COUPLINGS STIFF, AS = ALL FOUR COUPLINGS STIFF. EMG = ELECTROMYOGRAM, RA = RECTUS ABDOMINIS, ES = ERECTOR SPINAE, MVC = MAXIMUM VOLUNTARY CONTRACTION.

		No Exo	LC-NS	LC-LS	LC-AS	HC-NS	HC-LS	HC-AS
displacement peak (degrees)	F	2.4 ± 1.2	3.1 ± 0.7	2.9 ± 1.1	2.6 ± 1.0	2.1 ± 0.9	1.9 ± 0.9	3.0 ± 1.5
	B	3.2 ± 2.1	4.1 ± 2.0	4.3 ± 2.3	4.6 ± 1.0	4.2 ± 1.4	4.1 ± 1.5	5.1 ± 2.2
	S	2.0 ± 0.8	2.9 ± 2.3	3.1 ± 1.6	2.4 ± 0.3	2.1 ± 0.9	2.7 ± 1.7	3.0 ± 1.5
displacement 150 ms (degrees)	F	1.4 ± 0.8	1.6 ± 0.4	1.6 ± 0.5	1.5 ± 0.4	1.1 ± 0.7	0.9 ± 0.8	1.7 ± 0.6
	B	1.6 ± 0.8	1.5 ± 0.8	1.9 ± 0.6	1.7 ± 1.0	1.2 ± 1.0	1.0 ± 0.9	1.3 ± 1.0
	S	0.5 ± 0.6	0.7 ± 0.3	0.8 ± 0.8	0.8 ± 0.5	0.4 ± 0.4	0.6 ± 0.4	0.9 ± 0.5
peak right RA EMG (% MVC)	F	12.8 ± 8.9	16.8 ± 22.6	18.1 ± 27.8	19.8 ± 18.1	13.0 ± 13.7	6.4 ± 6.3	13.8 ± 8.9
	B	3.9 ± 1.8	3.9 ± 1.3	3.3 ± 1.7	3.7 ± 1.8	3.4 ± 1.5	4.6 ± 2.3	3.8 ± 1.6
	S	6.0 ± 5.4	2.6 ± 1.3	2.8 ± 1.1	3.7 ± 3.2	2.6 ± 1.6	4.2 ± 3.3	3.4 ± 2.1
peak left ES EMG (% MVC)	F	7.0 ± 2.1	5.4 ± 3.1	9.5 ± 3.6	8.2 ± 4.9	7.5 ± 3.9	10.1 ± 4.2	12.0 ± 8.7
	B	10.8 ± 6.7	6.3 ± 2.1	8.6 ± 7.1	10.0 ± 6.0	7.8 ± 5.7	10.2 ± 5.0	9.4 ± 4.5
	S	8.8 ± 2.8	4.9 ± 2.3	5.1 ± 2.2	6.0 ± 1.7	6.1 ± 3.0	6.7 ± 4.3	7.8 ± 1.9

IV. DISCUSSION

In our opinion, the effects of changing the exoskeleton's configuration are clearer in the second (perturbation) session, where there are noticeable differences in both trunk displacement and EMG. For example, for perturbations from the front, the combination of high compression and "lower stiff" stiffness reduces both peak trunk displacement (by 20%) and abdominal EMG (by 50%) while most other exoskeleton configurations actually increase both. For perturbations from the side, all configurations reduce EMG of both ES (by 10-45%) and RA (by 30-55%) muscles but largely increase displacement (by up to 50%); for perturbations from the back, nearly all configurations again reduce EMG (by up to 40% for ES) but have mixed effects on trunk displacement.

In the second session, there also appear to be interaction effects between stiffness and compression. For example, the peak trunk displacement when experiencing a perturbation from the front was lowest in the "high compression, lower couplings stiff" exoskeleton configuration, but both the "high compression, all couplings stiff" and "low compression, lower couplings stiff" resulted in higher displacement, and the decrease in the "high compression, lower couplings stiff" configuration thus cannot be clearly attributed to either compression or stiffness.

In the first session, all exoskeleton configurations increase peak flexion angle in all tasks compared to not wearing the exoskeleton (sometimes by over 50%). Furthermore, during the walking and box-lifting tasks, all configurations reduce peak low back extension moment compared to not wearing the exoskeleton. This may be thus simply an effect of the exoskeleton's base characteristics (e.g., basic stiffness, weight). While there are differences in both peak flexion angle and peak low back extension moment between exoskeleton configurations, they are not consistent between tasks. For example, high compression and fully stiff couplings result in the highest peak flexion angle during the sit-to-stand task but the lowest peak flexion angle during the box lifting tasks.

In the first session, peak EMG of the ES (a common outcome metric in lifting studies [2], [4], [6], [16]) is reduced for all exoskeleton configurations compared to no exoskeleton when lifting a 20-lb box (by up to 18%, similar to most passive exoskeletons [2], [16]), but not when lifting a 40-lb box, when especially high trunk compression actually increases peak EMG of the ES by up to 30%. There are no clear differences in the EMG of the RA, which is unsurprising since the tasks in the first session would not greatly activate the RA anyway.

A. Implications for Further Stiffness Evaluations

As mentioned in the Introduction, the 10% stiffness increase was not expected to have a major effect on participants; it was used to verify the study protocol and verify wearer confirm before increasing stiffness to a greater degree in a larger sample. Indeed, the ~10% increase in stiffness was largely not perceived by participants, who subjectively reported not feeling a difference between stiffness configurations. Furthermore, all participants tolerated the protocol with regard to length, intensity and exoskeleton comfort. Thus, in future work, we will induce larger changes in stiffness using the same protocol and systematically study their effects, more effectively characterizing the influence of stiffness in a trunk exoskeleton during different tasks. We believe that ideally stiffnesses of double or triple the original value should be investigated, and are exploring a compact spider spring and ratcheting mechanism to enable such large changes.

The changes in stiffness and compression should also be tailored to the wearer, similarly to how exoskeleton support and loads carried by participants are tailored to participants' characteristics in other trunk exoskeleton studies [2]. For example, a physically fit participant may not even notice changes in stiffness that may be constricting for a weaker participant. However, there is currently no clear guidance on how to tailor exoskeleton properties such as stiffness to the wearer, and we will explore this further in future research.

B. Implications for Trunk Exoskeletons in General

Results of this preliminary study indicate that trunk exoskeletons can be beneficial both for reducing muscle activity when lifting a box (session 1) and for resisting unexpected perturbations (session 2); however, the characteristics of the exoskeleton should be carefully taken into account to avoid negatively impacting the wearer. For example, in the first session, peak EMG of the ES is reduced by the exoskeleton when lifting a 20 lb box, but not when lifting a 40 lb box, and studies that evaluate an exoskeleton with only one lifted load (or more broadly, with only a limited subset of activities) may not obtain a complete picture of the exoskeleton's effects.

Furthermore, varying both compression and stiffness changes trunk displacement and EMG, and there appear to be interaction effects between stiffness and compression. Thus, different exoskeleton characteristics do need to be carefully considered during both design and deployment, and it is worthwhile to include either manual or automated adjustment of an exoskeleton to suit the current task.

However, given the small sample size, the results of the current study may be simply due to statistical noise. As a follow-up analysis, we examined individual data points to verify that the reported differences in means were not simply due to one large outlier. Nonetheless, in the future we plan to induce larger changes in stiffness with a larger sample.

V. Conclusion

A protocol for simultaneously varying both stiffness and trunk compression in an adjustable trunk exoskeleton was carried out with 6 participants, and found differences between exoskeleton configurations. While these differences must be taken with a grain of salt due to the small sample size, they indicate that both trunk compression as well as a relatively small increase in stiffness (which was not subjectively noticed by participants) may affect kinematics and EMG, that the effects may differ between activities, and that there may be interaction effects between stiffness and compression. In the future, we will conduct a larger study with the same protocol and a larger increase in stiffness in order to systematically characterize the effects of stiffness and compression in trunk exoskeletons, thus providing valuable knowledge about how such exoskeletons can be manually or automatically tailored to the wearer's behavior in order to optimize their effect.

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