A Backbone-Tracking Passive Exoskeleton to Reduce the Stress on the Low-Back: Proof of Concept Study*

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***Abstract*— Exoskeletons for the low-back have great potential as tools to both prevent low-back pain for healthy subjects and limit its impact for chronic patients. Here, we show a proof-of-concept evaluation of our low-back exoskeleton. Its peculiar feature is the backbone-tracking kinematic structure that allows tracking the motion of the human spine while bending the trunk. This mechanism is implemented with a rigid-yet-elongating structure that does not hinder nor constrain the motion of the wearer while providing assistance. In this work, we show the first prototype we manufactured. It is equipped with a traction spring to assist the wearer during trunk flexion/extension. Then, we report the results of a preliminary test with healthy subjects. We measured a reduction of the mean absolute value for some target muscles – including the erector spinae – when using the exoskeleton for payload manipulation tasks. This was achieved without affecting task performance, measured as task time and joints range of motion. We believe these preliminary results are encouraging, paving the way for a broader experimental campaign to evaluate our exoskeleton.**

I. INTRODUCTION

Many factors have sustained the recent interest on exoskeletons both in the industrial and healthcare sectors. Focusing on low-back exoskeletons, we can find the socio-economic impact of musculoskeletal disorders, such as low-back pain (LBP). Indeed, LBP counts as the leading cause of absence from work, especially considering the industrial sector. It is a serious health condition affecting a large part of the world's population. Its impact is dramatic: not only it causes absence from work [1], it is also the most common (and costly) cause of years lived with disability [2]. Nowadays, exoskeletons are exploited to target LBP with two complementary strategies: on the one hand, assistive exoskeletons may be used to restore lost functionality or mobility; on the other hand, occupational exoskeletons for the low-back are designed to reduce the stress on the musculoskeletal system, thus preventing or delaying the onset of low-back pain. The latter approach is recently being investigated and shows first signs of reduced acute physical stress and strain on the target area [3] and reduced local muscular activity [4]. Long-term effects should be studied in order to assess the prevention efficacy of occupational exoskeletons in reducing the occurrence and the impact of low-back pain. On the other hand, state-of-the-art rehabilitation for low-back pain still does not fully exploit exoskeletons or robotic technology. Indeed, postural rehabilitation [5], back school training (aimed at increasing awareness on back pain and proper movement execution) and focus groups are the most widespread approaches for chronic low-back pain [6]. Rehabilitation robotics and exoskeletons could improve the throughput of physical rehabilitation for LBP, mitigating both mobility and psychological limitations, and thus allowing a larger population to benefit of assisted physical therapy. In this work, we focus on occupational exoskeletons designed to assist the low-back. These devices can both assist impaired people – either with rehabilitation or work tasks – and support workers to prevent musculoskeletal disorders.

Occupational low-back exoskeletons are often grouped according to their actuation system. Passive exoskeletons rely on the deformation energy of elastic elements, such as springs or flexible beams. Active exoskeletons provide power mostly exploiting battery-powered brushless motors. Recently, hybrid solutions have been investigated, in which a low-power motor is combined with elastic elements to tune the provided assistance. Exoskeletons can be distinguished also considering their body interface [7]. Rigid exoskeletons exploit rigid frames to interface with the wearer and discharge their forces perpendicularly to the low back, avoiding additional compression of the spine. Soft exoskeletons (i.e., exosuits) interface with the body by means of soft garments. While they are generally more comfortable and less restrictive, they may provide additional compression forces, since they act in parallel with low-back muscles. Rigid exoskeletons are also preferred for occupational applications because of their higher output power.

In this contribution, we present our backbone-tracking rigid low-back exoskeleton. Its scissor hinge mechanism was previously described in [8]. Briefly, the exoskeleton features a backbone-tracking kinematic structure that allows the wearer to move as naturally as possible while being assisted. Thanks to this mechanism, the rigid structure of the exoskeleton elongates following the human spine.

In this work, we aim to design an experimental protocol to evaluate our low-back exoskeleton prototype. Hence, as a proof of concept, we have manufactured the prototype with the scissor hinge mechanism, and featured it with a spring to provide passive assistance. The device has been tested

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on two healthy volunteers during the execution of payload manipulation tasks, typical of the industrial sector.

The remaining of this contribution is structured as follows. In §II, we first show our low-back exoskeleton prototype, highlighting its backbone-tracking rigid structure, and then we describe our testing protocol. In §III, we show the results of our proof-of-concept study. Finally, in §IV, we comment on the results and draw some final remarks.

II. MATERIALS & METHODS

A. The Low-Back Exoskeleton

The human spine is a complex and articulated structure. Low-back exoskeletons are designed to relieve the lumbo-sacral joint (L5-S1). The exoskeleton should have anchor points on the body to properly assist the target region. Low-back exoskeletons usually have a rigid structure that is anchored at the level of the hips/legs and of the upper body. While bending forward the trunk, the human spine elongates as a result of trunk flexion. On the other hand, the rigid structure of an exoskeleton inevitably has a fixed length. This mismatch may result in constrained motion and discomfort for the wearer, as well as sub-optimal assistance.

The low-back exoskeleton prototype hereby tested features a backbone-tracking kinematic structure, as shown in Fig. 1. The scissor hinge mechanism is a relatively simple and scalable kinematic chain that enables the rigid exoskeleton to follow the motion of the spine. Its parametric design results from an optimization problem. Specifically, the structure has to minimize the tracking error of a reference vertebra (T2), acquired during squat and stoop lifting by means of infrared markers (cf. [8] for details).

We have equipped the exoskeleton with a spring that provides passive assistance. The spring is attached to the hip support body and engaged to a bar coupled with the mechanism (see Fig. 1). In this way, we achieve zero deformation of the spring when the wearer is standing, and thus no force is applied in this configuration. Trunk flexion causes the mechanism to elongate, tracking the motion of the spine, and the spring elongates in parallel. The springback produces an extension moment around the L5-S1 joint.

B. Test Protocol

The experimental data acquisition took place at the EuroBench facility [9] in Brunete (Madrid, Spain), after the approval of the local Ethical Committee. All subjects signed written informed consent. Two male healthy volunteers ($1.77 \pm 0.08\,\mathrm{m}$, $77.50 \pm 5.50\,\mathrm{kg}$) participated in the study. We wanted to evaluate the lift assistance capability of our low-back exoskeleton. In order to do this, each subject was tested in four conditions submitted in randomized order. These consisted in lifting either an empty cardboard box ($\mathrm{P} = 0$) or a 12-kg kettlebell ($\mathrm{P} = 1$), both with ($\mathrm{EXO} = 1$) and without ($\mathrm{EXO} = 0$) the exoskeleton. Subjects were instructed to perform stoop/semi-squat lifting. For each condition, the subjects completed 5 repetitions of the following experimental protocol: while standing in front of a table, lift the payload (either 0 or $12\,\mathrm{kg}$) from the ground and put it onto the table; wait 5 seconds; take the payload from the table and lower it to ground; wait 5 seconds. Fig. 2 shows one subject wearing the exoskeleton (left) and lifting the 12-kg payload (right).

Fig. 1. Low-back exoskeleton prototype with backbone-tracking kinematics (**a**) and passive actuation (**b**).

Fig. 2. Experimental setup and protocol: the subject manipulates a payload while wearing the exoskeleton.

C. Experimental Setup

We measured muscle activity and kinematic quantities during the experimental protocol. In particular, we used the Delsys Trigno wireless system (Delsys Inc., Boston, MA, USA). Each sensor has a bipolar electromyography (EMG) sensor and a 6-axis inertial measurement unit (IMU). EMG data was sampled at 2000 Hz and pre-filtered between 20 and 450 Hz, while IMU data was sampled at 148.148 Hz. We identified 6 muscles of interest and acquired a total of 12 EMG channels (left and right body segments). Specifically, we measured the muscular activity of the erector spinae (ES), the inferior rectus abdominis (RA), the semitendinosus (ST), the quadriceps femoris (QF), the soleus (SL), and the tibialis anterioris (TA). The skin was prepared by means of alcohol-wipe cleaning prior to attaching the EMG sensors.

D. Data Processing and Metrics Computation

The data acquired according to the experimental protocol were stored and analyzed offline using MATLAB R2020a

(The MathWorks Inc., Natick, MA, USA) and Python. We extracted the muscular activation signal from raw EMG data. First, we applied a zero-lag, band-pass (5^{th}-order low pass, 3^{rd}-order high pass) Butterworth filter to reduce the bandwidth of the signal between 30 and 450 Hz. Then, we applied full-wave rectification (FWR) and extracted the envelope with a moving average filter (75 coefficients, i.e., 37.5 ms). Finally, after FWR we normalized each signal with respect to the maximum value recorded among all conditions for that muscle. Given that the task described in §II-B is symmetrical (in the sagittal plane), we have computed the average activity as the mean of the left and right channels for each muscle of interest.

IMU data was filtered with a zero-lag, 2^{nd}-order low-pass Butterworth filter at 20 Hz. Then, we performed inter-sensor timestamp-based synchronization between EMG and IMU data, upsampling the IMU signals to 2000 Hz by means of linear interpolation. Finally, to reduce the dimensionality of the datasets, we downsampled both the inertial and muscular activity signals to 1000 Hz.

To evaluate the performance of the subjects and the effect of the exoskeleton, we computed both IMU- and EMG-based metrics and performance indicators. In particular, we exploited inertial data to measure the time to perform the task (i.e., task time) and the range of motion (RoM) for the hip and the knee. On the other hand, we evaluated muscle activity using the mean absolute value (MAV). We selected a subset of the 6 EMG signals corresponding to the following target muscles: quadriceps femoris (QF), semitendinosus (ST), erector spinae (ES), and rectus adbominis (RA). Intra-task actions, such as bending forward, lifting the payload, lowering the payload, and extending the trunk, were identified and segmented analyzing the inertial data measured by the sensors placed on the erector spinae. In particular, we analyzed the angular velocity signal in the sagittal plane. Data-driven thresholds were defined to detect the onset and offset of each movement, while the zero-crossing instant was used to detect the transition between trunk flexion and extension. The task time was then computed directly considering the so-identified time instants. In particular, we can define the *lifting time* (t_1) as the time interval that includes: bending the trunk to reach the payload on the ground and extended the trunk with the payload. Similarly, we define the *lowering time* (t_2) as the time interval that includes: bending the trunk to lower the payload, place it onto the ground, and fully extended the trunk. MAV values were computed within these time windows for trunk flexion and extension independently. Specifically, we computed the MAV only during the extension phase of lifting actions and the flexion phase of lowering actions (i.e., only during payload manipulation).

To compute the range of motion, we first estimated quaternions from raw IMU data. For this, we exploited the sensor fusion Kalman filter implemented by `imufilter` in MATLAB, based on [10], [11]. We used the IMU's embedded in the EMG sensors on erector spinae (ES), semitendinosus (ST), and soleus (SL). Then, we computed the Euler angles from the quaternions and considered only the rotation angle in the sagittal plane (denoted as α). The angle at the hip is then computed as the difference between α_{ES} and α_{ST}, while the angle at the knee is computed as the difference between α_{ST} and α_{SL}. Finally, we computed the range of motion as the maximum angular span for each repetition.

III. RESULTS

In this Section, we show the results of the tests carried out with two healthy volunteers wearing our low-back exoskeleton. Fig. 3 shows the mean absolute value (MAV) computed from the muscular activity signals of quadriceps femoris (QF), semitendinosus (ST), erector spinae (ES), and rectus abdominis (RA). The boxplots show the MAV computed for each repetition (colored dots) in each of the 4 analyzed conditions. We distinguished two actions within our experimental protocol: lifting (i.e., payload manipulated during trunk extension) and lowering (i.e., payload manipulated during trunk flexion). Hence, we have computed the MAV only during payload manipulation for each repetition of the protocol, and we show the results of trunk flexion and trunk extension separately. The exoskeleton has a negligible effect on the MAV of the QF for $\mathrm{P} = 0$ in the case of Subject 1 (S1), both for trunk extension and flexion. For $\mathrm{P} = 1$, instead, we notice increased MAV values. On the other hand, in the case of Subject 2 (S2), we notice different trend for extension and flexion. The exoskeleton reduces the MAV of the QF independently of the payload for trunk extension, although of different amounts. The opposite happens for trunk flexion. The trend for the ST is common between the two subjects. The exoskeletons has almost no effect on the MAV, that increases because of the different payload value. The activity of the erector spinae is affected by the exoskeleton for both subjects. On the other hand, we notice different trends for $\mathrm{P} = 0$. The MAV of the ES is decreased also in this case for S1, both for trunk extension and flexion, while it is increased for S2. Finally, we notice an increased RA activity independently of the exoskeleton for S1, both for trunk extension and flexion. In the case of S2, the exoskeleton reduces the MAV of the RA for trunk extension, and has a reduced effect on trunk flexion.

We evaluated the effect of the exoskeleton also considering some task-based metrics. Task time is shown in Fig. 4. Here we distinguish the lifting time (t_1) and the lowering time (t_2) (cf. §II-D). We notice a payload-dependent effect of the exoskeleton. For $\mathrm{P} = 0$, the exoskeleton increases the task time for both subjects. On the other hand, for $\mathrm{P} = 1$, the exoskeleton slightly increases t_1 and slightly reduces t_2 for S1, while it increases t_1 and has almost no effect on t_2 for S2. We also notice that t_2 is less affected by the exoskeleton in the latter case.

Finally, we show the results of the analysis on the range of motion for hip and knee in Fig. 5. We have considered lifting and lowering tasks separately, as for the task time. We can notice differences between the two subjects also in this case. On the one hand, the exoskeleton is reducing the RoM both at the hip and at the knee independently of the payload

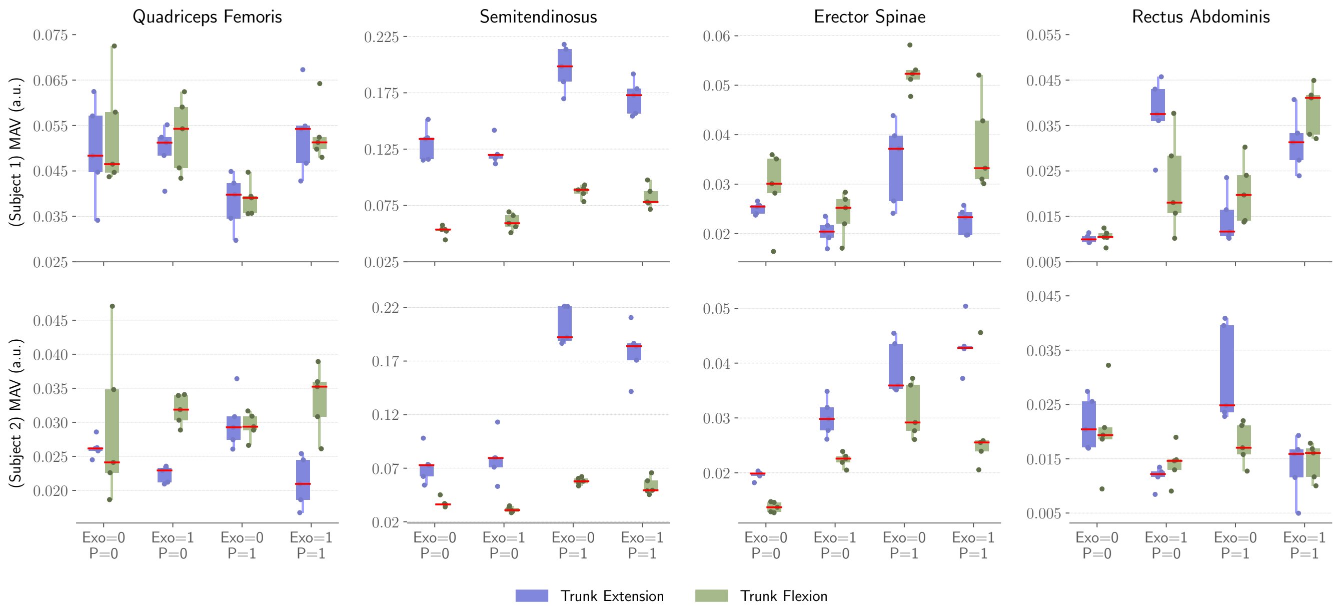

Fig. 3. MAV boxplot with dots of quadriceps femoris (QF), semitendinosus (ST), erector spinae (ES) and rectus abdominis (RA) for Subject 1 (top row) and Subject 2 (bottom row). Trunk extension and flexion are distinguished.

for S1. On the other hand, we notice the opposite trend for the RoM of the hip, and a reduced knee RoM only for $P = 1$ in the case of S2. With the exoskeleton ($EXO = 1$), hip and knee RoM's are comparable for the conditions $P = 0$ and $P = 1$, especially for the lifting task.

IV. DISCUSSION

In this work, we have shown and tested our low-back exoskeleton with a backbone-tracking kinematic mechanism. This is intended to follow the human spine while the trunk bends, removing a typical drawback of rigid-interface exoskeletons. To support trunk extension movements, we have equipped our prototype with a spring that provides passive actuation to the wearer and we performed some tasks typical of the industrial sector. In this scenario, low-back exoskeletons can either assist healthy workers in order to prevent low-back pain, or support workers with chronic low-back pain. We opted for a passive actuation system since it allows to evaluate the performance of the exoskeleton independently of the control strategy that would be necessary in case of an actuator. This proof-of-concept evaluation allowed us to validate the experimental setup and the testing protocol, paving the way for a broader evaluation campaign with a statistically-robust study population.

In §III, we have shown qualitative results that may help evaluating our low-back exoskeleton. In the following, we draw some remarks, considering also the limited size of the study population.

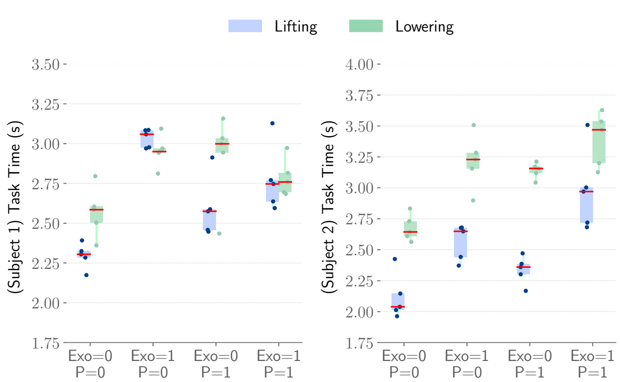

Fig. 4. Boxplot with dots showing the task time for subject 1 (left panel) and subject 2 (right panel).

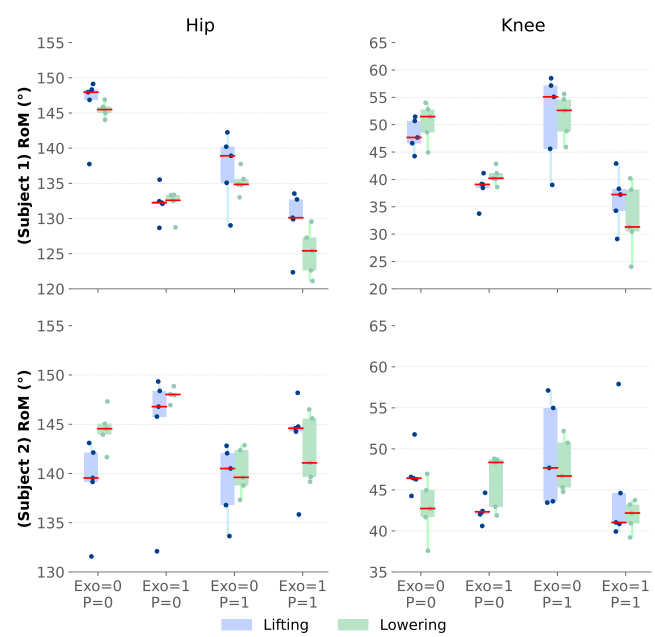

Fig. 5. Boxplot with dots showing RoM for hip (left panel) and knee (right panel) for subject 1 (top row) and subject 2 (bottom row).

There is also a worth-mentioning difference in both height ($1.85\,\mathrm{m}$ vs $1.70\,\mathrm{m}$) and weight ($83\,\mathrm{kg}$ vs $72\,\mathrm{kg}$) of the subjects. This difference may have impact on any exoskeleton. In fact, considering our prototype, it could also have effect on the backbone-tracking mechanism itself. As mentioned above (cf. §II-A), this design was obtained solving an optimization problem that allows to track the spine during trunk-bending tasks. The prototype we tested was first designed to fit Subject 1, and then featured with a height-adjustment mechanism to fit shorter/taller subjects. Sub-optimal wearer-exoskeleton matching could explain the different trends observed for the MAV of the muscles of interest, at least to some extent. Nevertheless, our exoskeleton seems to have the potential of reducing the activity of the muscles of interest. Specifically, the MAV for the erector spinae is reduced for both subjects for trunk flexion, and for S1 for trunk extension, while we notice a slight increase of the median value for S2.

It is fundamental to evaluate task-based metrics to understand the effect of the exoskeleton on end-user task performance. As shown above, the exoskeleton has increased the task time of payload lifting ($\mathrm{P} = 1$) for both subjects, although of different amounts. On the other hand, t_2 slightly decreased for S1 and increased for S2. Increasing the task time may be a positive effect of the exoskeleton, as it reduces sudden bursts of compression forces acting on the spine, thus reducing the stress on the musculoskeletal system.

Exoskeletons should not restrain nor limit user motion. For this reason, we have evaluated the range of motion of the hip and the knee. We noticed that the exoskeleton has some effect on both subjects, although different. For Subject 1, it is slightly reducing the RoM for both joints independently of the payload. We can also notice a reduction in hip RoM for S1 going from $\mathrm{P} = 0$ to $\mathrm{P} = 1$ without the exoskeleton ($\mathrm{EXO} = 0$). This may indicate subject-specific approaches to payload manipulation. We noticed a similar trend for the knee RoM of Subject 1. On the other hand, Subject 2 showed almost a dual approach to the task. The RoM of the hip is increased when using the exoskeleton independently of the payload. We can also appreciate how this subject used almost the same technique with ($\mathrm{P} = 1$) and without ($\mathrm{P} = 0$) the payload when not using the exoskeleton ($\mathrm{EXO} = 0$), differently from S1. Finally, the exoskeleton does not alter the kinematics of the knee for S2, especially for lifting. It is important to notice that the differences in range of motions do not alter task execution dramatically, as confirmed also by the aforementioned task-time results.

V. CONCLUSION & FUTURE WORK

In this work, we have shown our low-back exoskeleton as a proof of concept. Its purpose is to assist workers for trunk bending tasks, such as payload manipulation. Although we achieved some encouraging results, further research work needs to be done. Our backbone-tracking structure allows rigid-interface exoskeletons to track the motion of the wearer, with the aim of not hindering their motion while providing assistance to their musculoskeletal system, especially considering the low-back and the L5-S1 joint. As this preliminary analysis showed, we may need to improve the optimization-based design process of the mechanism. In particular, we plan to perform backbone tracking experiments with subjects of different BMI, to have different sizes of the exoskeleton with optimal fit. As the scissor hinge mechanism is the result of an optimization process, we can minimize the number of BMI-dependent parts to facilitate prototyping and manufacturing. The tested prototype was fitted with a non-optimized height-regulation system that probably induced a misalignment for subjects of different height. This could also be improved in order to *interpolate* the optimized mechanisms obtained from experimental data. In the near future, we plan to first re-design and rigorously evaluate the exoskeleton with passive actuation. Then, we would continue with the design and integration of an active actuation system and its control unit, to further improve the potential of our exoskeleton.

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