A low-back exoskeleton can reduce the erector spinae muscles activity during freestyle symmetrical load lifting tasks

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***Abstract* — Low-back wearable robots are emerging tools to provide support to operators during handling of goods and repetitive operations. In this paper, we present and validate a novel control strategy for an active pelvis orthosis, that operates intuitively and effectively to assist workers during lifting operations. The proposed control strategy has a hierarchical architecture: the first layer, the intention-detection, is deputed to the online detection of the onset of the lifting movement; the second layer, the assistive strategy, computes the reference torque profile to assist the movement, after the onset of the lifting movement is detected; the third layer, the low-level control layer, aims at setting the current to drive the actuators. The control strategy grounds on the sensor signals acquired by the robotic device and does not need additional sensors to detect the event. The system was tested on a pool of five healthy subjects requested to perform repetitive lifting movements: first, the subject was requested to lower the trunk, grasp the box, lift it up and place it on a table; second, the subject was requested to grasp the object from the table, lower it down, place it on the floor and get up without the load. The tasks were executed with the exoskeleton controlled in transparent and assistive modes. Results show that the assistive action allows to perform the lifting movement faster. Surface electromyography of low-back muscles show a reduction of the Lumbar Erector Spinae activity in the assistive mode compared to the transparent mode: a 16% reduction is observed when extending the trunk while holding the weight and a 33% reduction resulted when extending the trunk without holding the load.**

I. Introduction

Over the last few decades, automation has been spreading in the industrial world, due to the growing number of manufacturing companies requiring fast and precise production processes [1]. Despite the growth of automation, the sixth European Working Conditions Survey carried out in 2015 in 35 countries revealed that many workers are still exposed to some physical risks due to repetitive handling of goods and keeping tiring position at the workstation [2]. In last decades, to improve ergonomic conditions of workers, many companies implemented solutions with the final goal of improving the workplace, thus reducing the musculoskeletal diseases (MSD) and injury risks, based on the guidelines of the standard ISO 11228 and the NIOSH method [3]. Despite the improved workplace conditions, interventions do not seem sufficient yet, as work-related MSDs are still common and strongly affect blue collars, significantly impacting the related social costs. Among the MSDs, in 2010 data collected by the World Health Organization for the Global Burden of Disease revealed that low-back pain is still, overall, the most disabling condition for workers [4].

Recently, exoskeletons have been proposed as tools to provide support to operators during handling of goods and repetitive lifting tasks. For such application, exoskeletons are designed to assist the lumbosacral joint in order to reduce the effort required to the Erector Spinae muscles group for controlling the forward flexion of the trunk and its extension. These postural muscles typically bear the weight of the upper part of the body. When extending the trunk, these muscles should exert a force to accelerate the trunk upwards, which compresses the spine, possibly causing injuries of the soft tissues –muscles, ligaments and intervertebral disks– and can result in the occurrence of low-back disorders [5]. Recently, a review by de Looze and colleagues underlined the contribution that the exoskeleton technology could provide to assist workers [6]. Despite the reviewed exoskeletons were proved to lead to a reduction in the muscular activity, they found several issues that limit the current use of exoskeletons in factories, such as discomforts, weight of the devices, suitable physical robot-human interfaces, and intention detection algorithms to enable smooth assistance. Exoskeletons can be classified in passive and active devices, depending on their actuation strategies. Passive exoskeletons consist of springs and dampers to support a specific joint during movement: the assistance is provided only in one direction of the movement, while opposing to the other. Among passive exoskeletons for industrial application, the Personal Lift Augmentation Device (PLAD) has been largely validated and tested [7, 8]. Thanks to six springs aligned with the Erector Spinae and leg muscles, the PLAD assists the lumbosacral joint in trunk extension movements. Different studies have proved that the PLAD can reduce the back muscles effort (up to 40% with high level of spring stiffness [9]) and delay the muscular fatigue occurrence [10, 11], even though some subjects complained about discomfort and movement restrictions. On the other hand, active exoskeletons use external actuators to provide additional power to the user, in different directions. Among those, the Muscle Suit integrates pneumatic

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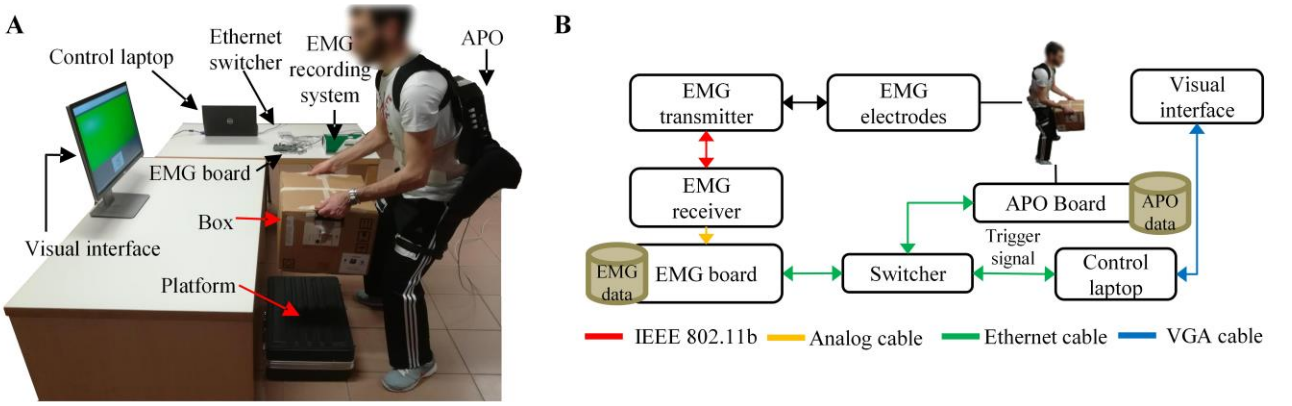

Figure 1. Experimental setup description. **A** A subject wearing the APO performs load lifting tasks when the LED on the visual interface turns on. **B** Block diagram describing the communication established among the devices used in the experimental activity. Light brown boxes depict where the collected data are stored.

actuators to assist the upper-limb and lumbosacral joints: in this case, the delivery of the assistance is manually triggered by means of an external switcher. Several load holding and carrying experiments showed a reduction of the Erector Spinae muscles activity up to 40% [12, 13].

Compared to passive exoskeletons, active devices show higher flexibility in the control, as they can provide torque along the whole movement. At the moment, the development of control strategies for providing intuitive and effective assistive strategies remains a challenge that has not been widely investigated. To develop an effective control strategy, two main design issues must be considered: first, the device should automatically detect the onset of the movement, in order to let the user relatively free to move when no lifting is detected, support the body weight and provide active assistance only when necessary; second, when a lifting movement is detected, the system should provide the assistive action smoothly and timely with the movement performed by the user, in a bioinspired way. Similar control requirements have been identified and implemented in lower-limb exoskeletons for gait assistance [14-16].

This work describes a novel control strategy for a low-back robotic exoskeleton, namely an Active Pelvis Orthosis (APO), to assist workers in lifting operations. The proposed control strategy has a hierarchical architecture: it can automatically detect the onset of the user's lifting movement and can deliver a fast yet-gentle extension torque to help the user's trunk extension. Five healthy subjects were recruited to validate the developed control strategy performing repetitive lifting tasks while measuring surface electromyography (sEMG) of relevant muscular groups.

II. Materials and Methods

This section describes the experimental setup and the developed control strategy, along with the experimental protocol and data analysis.

A. Active Pelvis Orthosis

The APO is a robotic hip exoskeleton for the assistance of the hip flexion/extension movement, developed at The BioRobotics Institute of Scuola Superiore Sant'Anna. Previous versions of the device were presented in [17, 18]. The mechanical structure of the APO included (i) a carbon-fiber frame structure connected to the user's trunk by means of an orthopedic shell and braces, and (ii) two rotating linkages connected at the user's thighs (Figure 1A). The axis of rotation of each of the two linkages is collocated with that of the human hip joint by means of two degrees of freedom: a passive hip adduction-abduction and the actuated hip flexion-extension. The actuation units employ a series elastic actuation (SEA) architecture [19], based on a custom torsional spring (different designs of the custom torsional spring have been presented in [20, 21]) and electromagnetic motors. Each SEA has a single-axis configuration composed of a motor-reduction stage connected to the torsional spring, whose deformation is directly measured by an absolute encoder. A second encoder placed on the APO hip axis measures the hip angle. In this study, we exploited the counteraction torque at the lumbar joint generated by hip extension torque, when performing lifting movements.

The control architecture of the APO is independent for each actuation unit and is based on a hierarchical architecture, i.e. a closed-loop feedback control setting the current to command the output joint torque delivered by the actuators, running on a field programmable gate array (FPGA) processor at 1 kHz, and the high-level control strategy, i.e. the algorithm providing the desired torque reference over the movement, running at 100 Hz on a NI System-on-Module real-time controller (National Instruments, Austin, Texas, US).

B. High-level control strategy

The control strategy consists of two main modules: (i) an *intention detection module*, which aims to automatically detect the onset of the trunk extension movement and trigger the assistive action and (ii) an *assistive strategy module* that computes the reference assistive torque, after that the onset of the movement is identified.

The intention detection module acquires the left and right hip joint angles and computes the average angle in real time. A rule-based algorithm detects the onset of the lifting movement if (i) the average hip angle overcomes a certain threshold, (ii) the variance of the average hip angle calculated in a time window of 10 samples is larger than a threshold and (iii) the average angle is decreasing. In addition, the end of the lifting movement is detected when (i) the average hip angle is lower than a threshold and (ii)

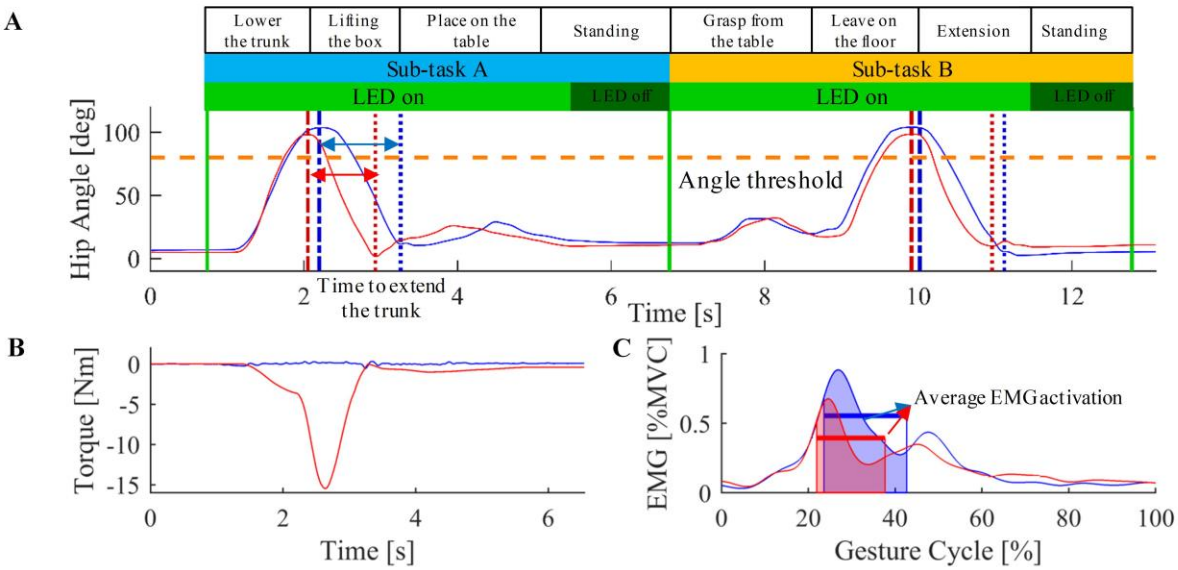

Figure 2. **A** Average hip angles profiles in transparent mode (in blue) and in assistive mode (in red) over the gesture cycle. The vertical solid light-green lines point at the time instants when the LED is on. For sake of simplicity, the LED turning-off is reported while the subject stands, because it is not related to any task accomplishment. Data segmentation is made between two consecutive light-green lines: the left profile shows the lifting, while the right one the lowering. The dashed orange line is the angle threshold set to the intention detection to detect the onset of the lifting. Vertical dashed lines are the initial and final moment of lifting detected by the intention detection module. The duration of the extension is computed as the difference between the two events. **B** Torque delivered by the exoskeleton in transparent mode (in blue) and assistive mode (in red) at both sides. Flexion torques are positive, extension torque negative. **C** Example of enveloped EMG signal and average muscle activity computation on the resampled signal. The blue signal is collected in transparent mode, while the red in assistive mode. The shaded areas depict the part of the EMG signal related to the trunk extension.

either its variability is smaller than a threshold or the average hip angle reaches a local minimum. The intention detection grounds on the assumption of symmetrical lifting; therefore, a single lifting movement is detected for both right and left sides.

The assistive strategy has been designed to generate a reference torque profile similar to the biological torque profile exerted by lumbar erector muscles by means of a patent-pending method [22]: a spring-like extension torque, proportional to the trunk flexion angle, is provided during trunk flexion movement, in order to support part of the body weight against gravity; moreover, an additional bell-shaped torque profile is provided after the detection of the onset of the lifting movement. The stiffness of the virtual spring for the assistance in the flexion phase and the amplitude of the bell-shaped torque in the extension phase are tunable control parameters, that can be selected based on the subjective perception of the assistance and comfort.

C. *EMG recording system*

To record the surface electromyographic signals, we used the TeleMyo 2400R EMG recording system (Noraxon Inc., AZ, USA). Pre-gelled bipolar Ag/AgCl surface electrodes (Pirrone & Co., Milan, Italy) were placed over the muscles following the SENIAM guidelines [23]. sEMG signals were collected unilaterally from five muscles: Lumbar Erector Spinae (LES), Thoracic Erector Spinae (TES), Erector Spinae Iliocostalis (ESI), Biceps Femoris (BF), and Rectus Femoris (RF). The former four muscles belong to the so-called posterior chain, i.e. the thoracic, lumbar and hip extensor muscles. The recorded signals were sampled at 1.5 kHz and then low-pass filtered at 500 Hz. The analog output of the EMG recording system was acquired at 1 kHz and stored by a dedicated board.

A UDP connection was established between the APO and the board reading the EMG signals from the EMG recording system for data synchronization.

D. *Experimental setup*

The experimental setup included a 10-kg box, a 90-cm table and a 20-cm platform. A computer screen placed on the table in front of the subjects displayed a visual LED for pacing the subject’s movements (Figure 1A).

The devices were connected to a control laptop by Ethernet cables through an Ethernet switcher (Figure 1B). On the control laptop, a dedicated routine ran for visualizing and saving data, controlling the exoskeleton and displaying the visual LED on the computer screen connected to the control laptop by means of a VGA cable.

E. *Experimental protocol*

Five healthy male subjects gave informed consent to participate to this study (age 29 ± 3 years, weight 69.6 ± 5.4 kg, height 174.6 ± 8.8 cm). Subjects were requested to wear comfortable sportswear and athletic shoes.

Upon arrival, an experienced experimenter placed the EMG electrodes on target muscles of the participant’s body. For each muscle, the subject performed two five-second maximum voluntary contractions (MVCs) against a resistance, with one minute of rest between consecutive trials, as done in [9]. After EMG placement and calibration, the subject worn the exoskeleton and was asked to familiarize with the system and the freestyle lifting task [24], in transparent mode (TM, i.e. the desired torque is set to 0 N·m) and assistive mode (AM, i.e. the desired torque is the output of the assistive strategy). In the assistive mode session, the control parameters were tuned based on the subjective feedback of the user, for a comfortable and effective assistance. Typically, the peak extension torque was set between 12 and 15 N·m. This session lasted around 20 minutes.

During the testing, the participant was requested to perform repetitive lifting movements, following a two sub-task procedure: first (namely, sub-task A), the subject was requested to lower the trunk, grasp the box from the platform, lift it up and place it on a table; second (namely, sub-task B), the subject was requested to grasp the object from the table, lower it down, place it on the platform and get up without the load. Each lifting and lowering action was triggered by the LED displayed on the screen; in particular, the subject was instructed to start to lower down when the LED turned on. Each lowering/lifting task was triggered every 6s. The lifting and lowering sequence was repeated 30 times. The trial was repeated both in TM and AM. A break of 10 minutes between the two trials allowed subjects to rest. The order of execution of the two trials was randomized across subjects.

F. Data analysis

Measured left and right hip joint angles and torques collected by the APO and EMG signals collected by the dedicated EMG board were analyzed offline by means of custom MATLAB™ routines (The Mathworks, Natick, USA).

The time to perform a trunk extension was estimated based on the classification of the intention detection module, i.e. when the algorithm recognized a lifting movement (Figure 2A).

Then, data of the two testing sessions were segmented into the two sub-tasks, namely the sub-task A and B, according to the visual cue displayed online (Figure 2A). Further segmentation was performed based on the events detected online by the intention detection algorithm, i.e. the onset and end of the lifting up movement (Figure 2A). APO and EMG collected data were resampled over the single on-off cycle of the visual cue.

The EMG envelope was obtained by applying to the raw data a high-pass filtering at 20 Hz, followed by rectification and low-pass filtering at 2 Hz (Figure 2C), using third-order Butterworth filters. The same processing was applied to EMG data of the MVC sessions. For each muscle, the maximum value between the two enveloped MVCs trials was used to normalize the correspondent enveloped EMG signal in the two experimental sessions, as done in [9]. The average EMG activity was computed on the enveloped signal, in the trunk extension phase, i.e. delimited by the events detected by the intention detection algorithm (1).

$$AvgEMG_{ext} = \frac{1}{T} \cdot \Delta t \cdot \sum_{i=1}^{N} X_i \qquad (1)$$

where X_i is the i_{th} sample of the signal, N is the number of samples in the epoch, Δt is the integration step and T is the time the subject took to perform a trunk extension. For each subject, the medians and interquartile range of the $AvgEMG_{ext}$ values were calculated for the two testing conditions. Statistical analysis was performed to evaluate whether the control strategy (namely the combination of the intention detection and the assistive torque) could reduce the muscular effort requested to the posterior chain muscles during the trunk extension movement. Features values were averaged across subjects and single tail paired-samples Wilcoxon signed rank test allowed to check for significant variations in muscular activity between the two testing conditions. Significance level was set at 5%.

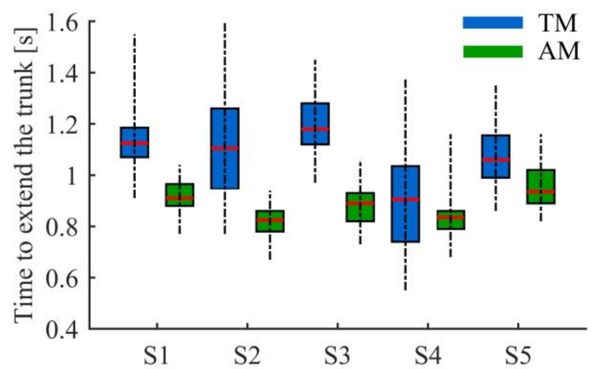

Figure 3. Time to extend the trunk in transparent mode (TM) and assistive mode (AM) among the five subjects. Boxes include data between the 25th and the 75th percentile, while the red lines are the median values. Dash-dotted lines are plotted between the minimum and maximum values of the time values.

III. RESULTS

A. Intention detection performance

The proposed intention detection module correctly recognized all the lifting movements performed by the subjects. No false negative or false positive events were detected throughout all the experimental sessions.

B. Time to extend the trunk

The time requested to perform trunk extension was significantly lower in the AM condition compared to the TM condition (Figure 3). The median reduction across subjects was –19.1% (–24.7% ÷ –10.8%) and it resulted statistically significant (single tail paired-samples Wilcoxon signed rank test, $p < 0.05$).

C. EMG activity

In sub-task A, across subjects, the $AvgEMG_{ext}$ values resulted in a variation between the TM and AM of –15.9% for the LES ($p = 0.06$), –6% for the TES ($p = 0.03$), –3.6% for the ESI ($p = 0.3$), –27.4% for the BF ($p = 0.06$) and +33.7% for the RF ($p = 0.09$).

In sub-task B, $AvgEMG_{ext}$ variations were: –33% for the LES ($p = 0.03$), –10.1% for the TES ($p = 0.03$), –8.9% for the ESI ($p = 0.4$), +7.1% for the BF ($p = 0.15$), +40.1% for the RF ($p = 0.03$).

Results of LES, TES and ESI are reported in details in Figure 4 for the five subjects.

IV. DISCUSSION

This paper presents a novel control strategy for a robotic low-back exoskeleton that supports users while lifting goods. The system provided left and right hip extension assistance, with an average torque peak of 15 N·m; therefore, the counteraction torque exerted at the trunk, around the lumbosacral joint, resulted in about 30 N·m. This

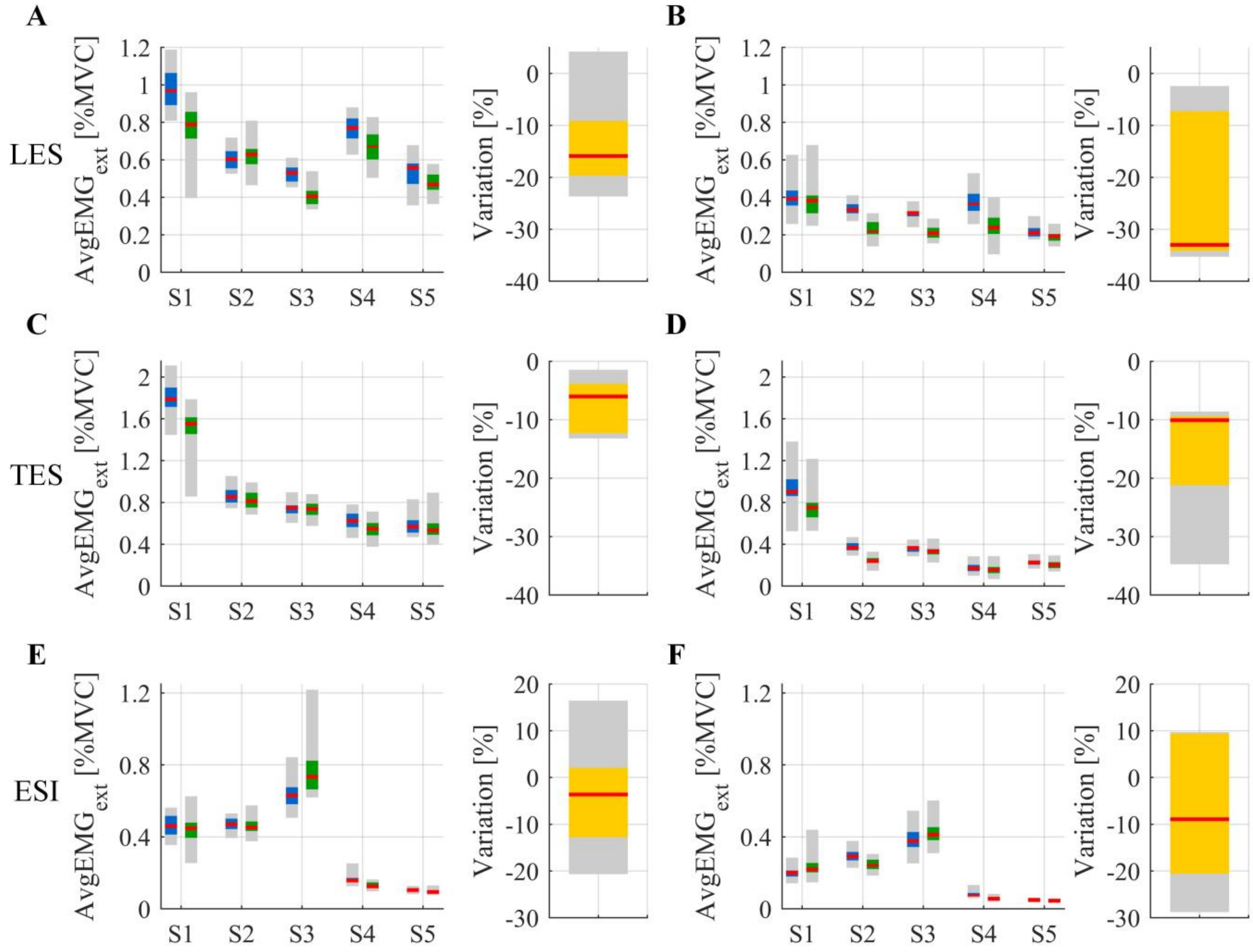

Figure 4. Comparison of the $AvgEMG_{ext}$ values of the three back muscles computed on the enveloped signal over the trunk extension, for the five subjects between transparent mode (in blue) and assistive mode (in green). Each panel reports on the right side the achieved percentage variation among subjects. Red lines are the median values; colored boxes include the data between the 25th and the 75th percentile; grey boxes include the data between the minimum and the maximum values. **A** $AvgEMG_{ext}$ values and variation of the LES muscle in sub-task A. **B** $AvgEMG_{ext}$ values and variation of the LES muscle in sub-task B. **C** $AvgEMG_{ext}$ values and variation of the TES muscle in sub-task A. **D** $AvgEMG_{ext}$ values and variation of the TES muscle in sub-task B. **E** $AvgEMG_{ext}$ values and variation of the ESI muscle in sub-task A. **F** $AvgEMG_{ext}$ values and variation of the ESI muscle in sub-task B.

amount of torque is about 15% of the biological torque developed at the lumbosacral joint when lifting 12.5 kg [5]. In literature, some exoskeletons provide the same amount of torque [25], while the Muscle Suit can deliver 90 N·m, thanks to an external compressor which supplies the onboard pneumatic actuators [13].

The intention detection module correctly recognized all the lifting movements performed by the subjects, showing high reliability in delivering the assistive torque. Compared to many state-of-the-art methods, which used external buttons or joysticks to trigger the exertion of the assistance [26, 27], the approach proposed in this paper allows the automatic triggering of the assistance, with no need of additional sensory information (such as from EMG or IMUs signals placed on the user's body) and does not require any action from the user. Thus, this strategy can intuitively detect relevant events and significantly improve the acceptability by final end-users in target scenarios, such as production lines or warehouses.

By analyzing the average EMG values over the trunk extension, we observed a general reduction of the muscle activity in all posterior chain muscles in AM condition with respect to TM condition. In most cases, the reduction achieved the statistical significance. When statistical significance was not achieved, we observed a p value close to the significance level, suggesting the existence of a trend in the collected data. It is fair to notice that for the computation of the $AvgEMG_{ext}$ we relied on the accuracy of the intention detection module in timely recognizing the onset and the end of the trunk extension movement. Therefore, the segmentation of the EMG signal could be affected by the performance of the intention detection module. Nevertheless, we believed that focusing our analysis on the extension phase of the movement might lead to a better understanding of the effects of the delivered assistance to the Erector Spinae muscles, rather than performing the computation over the whole movement which could include information not related to the lifting movement itself (e.g. movements to maintain balance).

In literature, a study about the Muscle Suit quantified the effects of the exoskeleton assistance by comparing the integral of the EMG signal while extending the trunk in AM to the condition without the exoskeleton [27]. In that case, the integral was not normalized by T. They found a reduction of the erector muscles activity by 40% while extending the trunk without load with the Muscle Suit with respect to extending the trunk without both the load and the device; instead, they observed any variation while lifting a 10-kg load with the Muscle Suit with respect to extending the trunk without a load with the Muscle Suit. Interestingly, when we computed the integral of the EMG signal on our data, we achieved a reduction of the 19.8% and 17.6% respectively for the LES and the TES when lifting a load (both $p < 0.05$) and a reduction of 38.8% and 34.7% for the same muscles when extending the trunk without a load (both $p < 0.05$). However, our comparison is with respect to the TM condition.

Notably, the not-significant variation observed in the ESI activity could be related to the constraints introduced by the APO lumbar orthotic shells, which could prevent the user from performing asymmetrical trunk movements and may not be affected by the assistive action provided by the APO.

Similar to other studies [6], the RF $AvgEMG_{ext}$ values increased when extending the trunk in AM condition. This result might be caused by an involuntary muscle co-contraction due to the assistance to help maintain balance.

The APO was able to effectively support the load lifting task providing an assistive action that allows to perform faster lifting but requiring less muscular effort. Future studies will compare the AM condition to an ecological condition, i.e. in which the task is executed without wearing the exoskeleton. Moreover, further studies will be performed to assess the accuracy of the intention detection algorithm in the timely detection of the lifting movement.

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