Rigid, Soft, Passive, and Active: a Hybrid Occupational Exoskeleton for bimanual multijoint assistance

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***Abstract*—Physically demanding work is still common in western countries, with large proportions of the workforce that are exposed for more than a quarter of their working time to tiring postures or repetitive tasks: the shoulder is one of the main body areas susceptible to work-related musculo-skeletal disorders. Recent advancements in assistive technology have provided new instruments to promote safety and reduce workload. Colloquially referred to as occupational exoskeletons (OE), these wearable devices are usually spring-loaded, and provide gravity support for overhead tasks. OEs for upper limbs are usually single-joint exoskeletons and assist shoulder flexion/extension; they do not provide support to distal joints such as the elbow. In the present work, starting from a commercially available exoskeleton, we propose an innovative concept of hybrid upper-limb OEs. We combined a spring-loaded shoulder exoskeleton with an active elbow exosuit to extend the capability of the OEs to provide gravitational support to both shoulder and elbow flexion-extension in strenuous manual tasks. The proposed device can reduce up to 32% of the *biceps* activity during the elbow flexion and up to 31% of the *deltoids* activity during the shoulder abduction. In-lab experimentation showed the potentials of such a hybrid approach in reducing the strain of the upper-limb muscles.**

***Index Terms*—Occupational Exoskeletons; Exosuits; Embedded Control; Assistive Devices.**

Fig. 1. *Hybrid bimanual exoskeleton* developed to assist the elbow and shoulder joints; on the top-left the front view of the wearable device with positioning of the principal components, on the bottom-right back view of the suit with attention drawn to the actuation stage and control stage.

I. INTRODUCTION

Work-related musculo-skeletal disorders (WMSDs) cover 60% of work-related injuries [1] and are the most prominent health problem in the European Union (EU), affecting millions of workers and impacting the healthcare system with avoidable financial burdens [2]. In 2015, approximately three out of every five workers in the EU reported WMSD in the back, upper limbs and/or lower limbs [3] with a preponderance of backache and muscular pains in the upper limbs (43% and 41%, respectively), followed by complications in the lower limbs (29%) [4]. Focusing only on the upper limbs, it has been demonstrated that working with the hands above the shoulder level for more than one hour per day can result in several complications, like subacromial impingement syndrome, tearing of the rotator cuff muscles, as well as supra-scapular nerve compression [5].

Occupational exoskeletons (OEs) are being developed with the aim to make work safer by reducing the effort required of the back muscles during lifting, or the effort of the shoulder muscles during tasks of overhead manipulation [6]; from the beginning of the first concepts of exoskeletons, a variety of devices using different materials and designs have been proposed [7], to help people in different tasks and environments [8]. Upper-limb devices for supporting tasks of overhead manipulation, either static and dynamic, are designed commonly using rigid (or semi-rigid) structures (MATE from

Manuscript received: September, 8^{th}, 2021; Revised November, 29^{th}, 2021; Accepted December, 30^{th}, 2021.

This paper was recommended for publication by Editor Jee-Hwan Ryu upon evaluation of the Associate Editor and Reviewers' comments.

We acknowledge Sporlastic GmbH for providing us with the materials to build the exosuit and MathWorks for the technical support. We also acknowledge IUVO and COMAU providing us with the free loan of the MATE; N.V. and S.C. have commercial interests in the MATE technology being shareholders and advisors of IUVO S.r.l., a company owning the IP of the MATE. This work was supported by the project HeiAge and SMART-AGE (Project No.: P2019-01-003) by Carl Zeiss Foundation.

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Digital Object Identifier (DOI): see top of this page.

COMAU [9], ShoulderX from SuitX [10] and PAEXO from Ottobock [11]).

In recent years, there has been an increasing amount of literature documenting the actual effectiveness of OEs, with particular attention to the impact of exoskeletons on human biomechanics, across a large variety of work environments and specific manual tasks [6], [12]. As an example, in their recent paper, Maurice et al., performed an in-lab assessment of an upper-limb passive exoskeleton during a repeated overhead task in a group of twelve subjects [11]. They demonstrated that the device was able to reduce shoulder physical strain as well as global physiological strain, without significantly increasing lower-back strain nor degrading balance. A similar study from Schmalz et al. investigates the effects of the same device by analyzing the electromyographic (EMG) signals of different shoulder and back muscles as well as kinematic and metabolic parameters[13]. It was found that all evaluated muscles showed reduced activity when the device was used, accompanied by a reduction in both heart rate and oxygen rate. A common feature of the vast majority of commercially-available spring-loaded upper-limb OEs is that their assistance targets the shoulder elevation movement, without supporting more distal joints such as the elbow. On one hand, the lack of assistance for other joints is the consequence of a design strategy that aims at developing devices extremely light-weighted devices (close or lower than $3\,\mathrm{kg}$) to promote the highest level of acceptability and usability.On the other hand, the absence of assistance at the elbow restricts the type of tasks in which the exoskeleton can be beneficial to those activities that require the arms elevated, whereas several work activities typically involve complex gestures that could benefit from synergic support at the shoulder and the elbow [14]. Indeed, in tasks where the operators handle tools (e.g. varnishing, maintenance, logistics), providing gravitational assistance to the elbow could further reduce the muscle strain as well as the global fatigue, with potentially a more significant impact on the ergonomics risk. Unfortunately, the inclusion of a passive system to support elbow motions significantly affects the OE's overall weight, and the assistance modulation needs also to consider the shoulder kinematics. Alternatively, an actuated mechanism to assist the distal joint would add weight distally, increasing the inertia of the moving part and the overall system complexity [15], or, in the case of exoskeletons that unload their weight from the user through a rigid connection to the ground, would limit the wearer's mobility [16].

In this work, we propose a new combination of devices that derives from the current concept of OEs for overhead tasks with the ultimate goal to feature upper-limb OEs with elbow gravitational support. Even if there are already active wearable devices that assist simultaneously the elbow and shoulder [17], such a previous approach was based on vocal control with a predefined kinematic motion assistance, therefore limiting the dexterity of the wearer and restricting the use of the exoskeleton to a single specific lifting task. In our view, assisting elbow joint motions via the inclusion of a soft exosuit could be a promising direction to add functionality to the existing shoulder OEs, while limiting the impact on the inertia of the moving part, as well as the overall weight and complexity of the system. The idea of providing a hybrid assistive support (passive & active) and via a hybrid architecture (rigid & soft) is disruptively novel in the field of wearable robotics and, to the best of our knowledge, not yet proposed in the literature. More specifically, we aim at demonstrating that the proposed hybrid device, employing both active and passive actuation strategies, can be used to support intuitively multiple joints involved in the task without hampering physiological motion, despite the different working principles of the support. We started from previously published works [18] which highlighted abnormal co-contraction strategies in presence of single joint assistive passive devices and we investigated if such phenomenon was still persisting when multijoint assistance were provided. We investigated if the device was able not only to reduce the EMG activity on the muscle groups insisting on the shoulder and elbow, but also to allow physiological motion and preservation of muscle activation patterns.

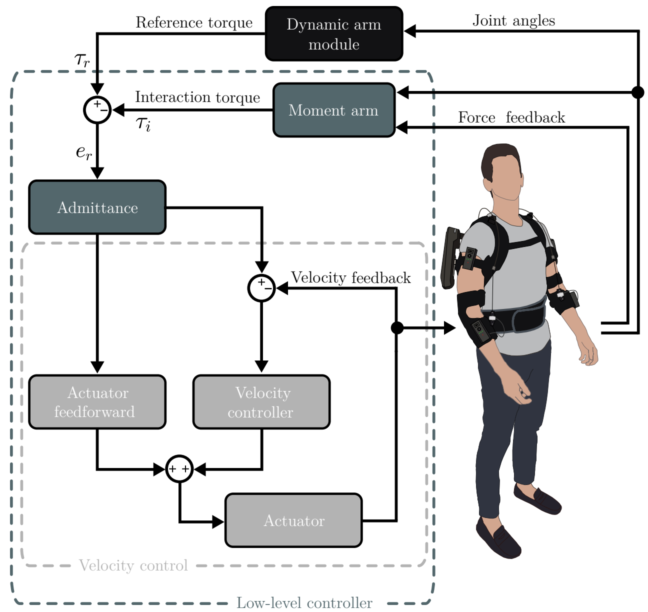

Fig. 2. Real-time control framework: The high-level controller is based on an inverse dynamic reconstruction of wearer's arms. We estimated a reference torque $\tau_r(t)$ from a biomechanical model, accounting for the tridimensional kinematics of the joints. This is then tracked by an admittance controller comprised of an outer torque loop (dark grey) and an inner velocity loop (light grey). The estimated torque is compared to the torque delivered by the exosuit to the human joint (τ_i). The error torque (e_r) is then converted into a motor velocity, delivering assistive power to the suit's wearer.

II. Methods

A. Hybrid exoskeleton design

The proposed hybrid device, shown in Figure 1, is a fully embedded system comprising of two modular and interconnected layers that can be worn independently: a new prototype of actuated soft wearable exosuit that supports the elbow flexion, and a passive OE assisting the shoulders in overhead static and dynamic tasks (i.e. *MATE*, COMAU, Turin, Italy). The assembly of the two modules has an overall weight of $6.4\,\mathrm{kg}$ and it is powered by a single battery pack (Tattu, $14.8\,\mathrm{V}$, $3700\,\mathrm{mAh}$, 45C).

Two passive actuation units store and transform the elastic energy of two parallel springs into assistive action to support

Fig. 3. *Experimental Setup*: the control stage streamed the real-time IMUs data via serial protocol to the visual interface on the screen, giving a visual feedback of the user's motion while following the trajectory of the dummy; in the figure, the user wore the hybrid exoskeleton and the two weighted bracelets.

the shoulder movements. This mechanism generates a resulting torque that varies consistently with the gravity torque profile of the arm in a physiological shoulder elevation [19].

The active layer (soft elbow exosuit) consists of a textile harness connecting arm and forearm, opportunely modified from a passive orthosis (Sporlastic Neurolux II, Nürtingen, Germany). The exosuit, partly integrated in the OE supporting structure, required an ad-hoc redesign of sensing, actuation and wearability respect to our previous architecture to obtain a fully embedded device [20], preserving the wearability of the MATE. The actuation stage supporting the elbow flexion, is mirrored for the right and left arms of the user, and it consists of two flat brushless motors (T-Motor, AK60-6, $24\,\mathrm{V}$, 6:1 planetary gear-head reduction, Cube Mars actuator, T-MOTOR, Nanchang, Jiangxi, China). The two motors drive two pulleys ($\varnothing 35\,\mathrm{mm}$) around which the artificial tendons are wound (Black Braided Kevlar Fiber, KT5703-06, $2.2\,\mathrm{kN}$ max load, Loma Linda CA, USA). Elbow assistance from the actuation stage to the exosuit is obtained via a Bowden cable (Shimano SLR, $\varnothing 5\,\mathrm{mm}$, Sakai, Ōsaka, Japan), connecting the motor pulleys to the anchor points placed on the textile harness (Figure 1) and located on the proximal and distal sides of the elbow joints (see multimedia attachment). Each artificial tendon is equipped with a force sensor (ZNLBM-1, $20\,\mathrm{kg}$ max load, Bengbu Zhongnuo Sensor, China) connected to the distal anchor point, which measures the interaction between the exosuit and the user's arm. The control architecture, and the power supply are encased in a 3D printed structure placed on the lumbar support of the passive exoskeleton (*MATE*).

The sensing network for motion detection and closed-loop control of the two elbow exosuits comprises four IMUs, two for each arm (IMU, Bosch, BNO055, Gerlingen, Germany), and two microcontrollers (Feather nRF52 Bluefruit, Adafruit Industries, New York City, USA) which allow the sensors to communicate with the actuation control unit positioned on the back of the device. Each IMU sensor communicates via Feather the quaternions corresponding to the arm posture and the load cell detecting cable tension via a proprietary Bluetooth Low Energy serial protocol (BLE UART, Nordic Semiconductors, Trondheim, Norway) to the other Feather board placed on the control stage. The receiver unpack the buffered data coming from the sensing units placed on the user's arms and communicates them via I2C with an Arduino MKR 1010 WiFi (Arduino, Ivrea, Italy), responsible for the real-time control unit. The Arduino board, running at $100\,\mathrm{Hz}$, is responsible of running the real-time controller, implemented in a MATLAB/Simulink application (MathWorks, Natick, Massachusetts MA, USA), and of sending the motor command via CAN-bus to the actuation stage.

B. *Real-time control framework*

The active layer, providing assistance at the elbow, controls the exosuit elbow flexion by means of an inverse dynamics approach (Figure 2). The *high-level control framework* estimates, in real-time, the human torque at the elbow joint, using kinematic information collected from the IMUs. The *low-level control framework* tracks such gravitational torque using an indirect admittance-based force controller.

1) High-level control framework: The high-level control framework reconstructs the user elbow torque by using a biomechanical model scaled on the user's anthropometrics. It estimates the reference torque $\tau_r(t)$ accounting for the tridimensional arm postures (i.e. elbow and shoulder), extracted in real time from the IMUs quaternions. The reference torque includes inertial, centrifugal, Coriolis and gravitational forces input in the equation of motion to compute torque at the joints τ^h and used the component at the elbow to close the loop of the controller:

$$\tau^h(q(t)) = M(q(t))\ddot{q}(t) + C(q(t),\dot{q}(t)) + G(q(t)), \qquad (1)$$

where q, $\dot{q}$ and $\ddot{q}$ represent the vectors of elbow and shoulder position, velocity, and acceleration. The inertial properties of the arm are described by the matrix $M(q(t))$; $C(q(t),\dot{q}(t))$ is the vector of Coriolis and centrifugal forces and $G(q(t))$ is the vector of gravitational force, extracted from the Opensim musculoskeletal model [21] and scaled on the subject's anthropometry. The estimated human biomechanical torque τ^h is a vector that comprises four elements: the first three elements represent the shoulder joint torques (i.e. shoulder flexion/extension, shoulder abduction/adduction, and shoulder internal/external rotations), while the last element is the elbow flexion/extension torque, which corresponds to the dynamic arm module reference torque τ_r.

2) Low-level control framework: The reference torque $\tau_r(t)$, estimated from the high-level controller, is used as input signal for the low-level admittance controller, which compares $\tau_r(t)$ with the interaction torque, $\tau_i(t)$ estimated from the force sensor recording the tendon tension, following the procedure described in [20]. The resulting instantaneous torque tracking error $e_r = \tau_r - \tau_i$ is transformed into a desired angular velocity, ω_r, through a PID-like admittance block of the form:

$$Y(s) = \frac{\omega_r}{e_r} = \frac{K_p + K_i \cdot s^{-1}}{1 + K_d \cdot s} \qquad (2)$$

where K_p, K_i and K_d gains were experimentally tuned, using the Ziegler-Nichols heuristic method, prior initiation of the study and left unchanged for all the participants [22].

C. Experimental setup

A DAQ board (Quanser QPIDe, Markham, Ontario, Canada) was used to acquire the data from the control unit via serial protocol, with a sampling frequency of $1\,\mathrm{kHz}$. Data from the control unit were used to provide real-time instructions to the subjects through a graphical interface (GUI) during the test (Figure 3). We used a multi-channel and wireless surface EMG system (Delsys Trigno, Natick MA, USA) to monitor six muscles on each arm of the subject (details can be found in the multimedia material): the long head of biceps brachii, the lateral head of triceps brachii, anterior, medial and posterior parts of the deltoid and the upper trapezius; electrodes placement followed the SENIAM guidelines, and we ensured that they were not colliding with the straps or the cuff of the device. Moreover, a signals quality check has been performed after each experimental session in order to verify the presence of significant artifacts in the muscular activity (i.e. electrodes detachment, movement artifacts, collision with the device). The EMG data and the exoskeleton data were synchronised offline using an external trigger signal, as EMG data was not used for control purposes.

D. Participants

Six healthy male participants were enrolled in the experiment (age 28 ± 8.51 years, mean $\pm$ SD, body weight $91.83{\pm}8.93\,\mathrm{kg}$ and height $1.89{\pm}0.03\,\mathrm{m}$). Inclusion criteria were: anthropometric sizes of the participants within the ranges accommodated by the shoulder and elbow modules, no evidence or known history of musculoskeletal or neurological diseases, and normal joint range of motion and muscle strength. All subjects provided explicit written consent to participate in the study. All experimental procedures were carried out in accordance with the Declaration of Helsinki on research involving human subjects and were approved by the IRB of Heidelberg University (Nr. S-311/2020).

E. Experimental protocol

We designed the experiment in order to evaluate subjects' performance when wearing the hybrid device and determine if the physiological movement is altered. Throughout the experiment, the subjects were equipped with the IMUs to record arms kinematics and with a load of $1.5\,\mathrm{kg}$ on each wrist to simulate the use of a working tool (e.g. a drill). The protocol consisted of a set of movements (tracking task) that were performed in two different conditions, namely, wearing the hybrid device that provided passive support to the shoulders and active assistance to the elbows (*Exo* condition), and wearing only the IMUs to record arms kinematics during the experiment (*No Exo* condition) without donning the hybrid device. Subjects were requested to perform repetitive bilateral motions, following the movements shown in the GUI by a phantom avatar. To allow the subject to track the avatar's movement, the GUI also showed the real-time representation of the subject's limbs overlapped with the avatar's limbs, the latter set with a transparency level of 30%, as depicted in Figure 3. The trajectory followed by the participants comprises a series of 12 elbow flexion-extension movements per trial with amplitude of $30°$, $60°$ and $90°$, combined with a sequence of 12 shoulder abduction-adduction movements per trial with amplitude of $35°$, $70°$ and $105°$. The joints movements have been combined in such a way to: (i) record both single and multijoint trajectories, (ii) study the influence of the assistance on physiological kinematics, and (iii) replicate the speed profiles of a previously published contribution [23], with an angular speed of $35°/\mathrm{s}$. The two trajectories were merged together after randomising the combination of the shoulder and elbow angles. The order of the tasks and conditions was also randomised to avoid biased behaviours. To avoid fatigue, subjects rested for $10\,\mathrm{min}$ between conditions. Before initiation of the experiment, a familiarisation phase allowed the participants to experience the assistance of the hybrid device.

F. Data analysis

The performance of the device was quantified in terms of its effect on muscular activity and movement kinematics. To assess the tracking accuracy, for each task, we computed the coefficient of determination r^2 and the Root Mean Square Error (RMSE) between reference and measured elbow and shoulder trajectories performed by the subjects over the two different conditions (*Exo* and *No Exo*). In addition, to determine the difference in the user's kinematics between the two conditions, we evaluated the delay of the movements compared to the displayed track by computing the cross-covariance between the signals. Before starting the experiments we collected maximum voluntary contraction (MVC) from the muscles recorded and used it for the EMG signal normalization during the data processing. Raw EMG signals were acquired at $2.2\,\mathrm{kHz}$ with the proprietary software of the wireless surface EMG system, EMGWorks (Delsys Trigno, Natick MA, USA); we used an external analog trigger to synchronize the EMG signals and the data recorded on the DAQ board during the offline signals processing. The EMG signals were filtered offline with a fourth order Butterworth filter (cut-off frequency $15\,\mathrm{Hz}$-$450\,\mathrm{Hz}$), rectified and low-pass filtered at $6\,\mathrm{Hz}$ with a fourth-order Butterworth filter; the signals were then normalized to each participant's MVC. We used the EMG signals as a metric to quantify the effort that the user required to the musculoskeletal system. Analysis of the muscular activity was performed by comparing the traces of the six muscles per arm in the phases of flexion and extension; raw EMG signals were processed offline to evaluate their Root Mean Square (RMS) as an index of activation level across tasks and conditions. The flexion and extension phases of the elbow and shoulder joints have been identified with a velocity threshold-based method [24]: first, the angle trajectories were filtered with a Savitzky-Golay filter, then the angular velocity was computed and segmented considering the 10% of velocity peak to identify the onset and offset events.

Furthermore, we assessed the Co-Contraction Index (CCI) related to the elbow and shoulder joints according to the formula:

$$CCI(\%) = 2 \times \frac{\int min(EMG_f(t), EMG_e(t))dt}{\int (EMG_f(t)dt + EMG_e(t)dt)} \times 100 \quad (3)$$

where the numerator of the equation reflects the amount of overlapping activity from both muscles EMG_f (flexor muscles, i.e. *biceps* and *anterior deltoids*) and EMG_e (extensor muscles, i.e.*triceps* and *posterior deltoid*), divided by the sum of both areas.

G. Statistical analysis

Data normality distribution was assessed using Shapiro-Wilk test. Performance indexes were compared across conditions with a two-way ANOVA; the analysis was performed with MiniTab (Minitab, State College, PA, USA). We evaluated the two conditions (*No Exo* and *Exo*) as first factor and the two arms of the subject as second one. In the EMG analysis we assessed the difference of the six muscles recorded in the elbow and in the shoulder rise and fall phases (elbow flexion/extension and shoulder abduction/adduction), and their mutual interaction. We evaluated also the difference related to the users' kinematics (coefficient of determination r^2, Root Mean Square Error (RMSE) and delay). When the ANOVA results were significant, we performed a post-hoc analysis applying the Fisher's LSD test to evaluate the significant pairwise differences between each type of assistance based on the distribution of the data. Statistical results for ANOVA follow the standard notation F(n, d), where n are the DOFs of the numerator (i.e. No Exo, Exo) and d of the denominator (i.e., subjects). For all the tests, the level of statistical significance was set to 0.05. Reported values and measurements are presented as mean ±standard error (SE). We highlighted significant differences in the results with the symbol * in all the figures.

III. Results

A. Tracking accuracy

We report the position tracking accuracy in Figure 4 for both elbow and shoulder joints. Figure 4a shows the elbow trajectories for a representative subject during the tracking task in the two analysed conditions (*No Exo*, *Exo*). We report values averaged across the right and left arm, as the two-way ANOVA showed no arm-related difference in the subject group, at the population level, between the two limbs ($F_{1,5} = 0.01$, $p = 0.929$). However, we see significant differences ($F_{1,5} = 4781$, $p < 0.0005$) in the coefficient of determination r^2 between the desired and measured trajectories, across the two conditions (*No Exo* and *Exo*) (0.87±0.006, 0.73 ± 0.013). This is confirmed by the RMSE ($F_{1,5} = 19.45$, $p < 0.0005$), where we observe an error of $12.72 \pm 0.78°$ in *No Exo*, and of $21.03 \pm 0.86°$ in the *Exo* condition (Figure 4c). At the shoulder level (Figure 4b and 4d), we also find significant difference ($F_{1,5} = 8.26$, $p = 0.009$) in the r^2 between the *No Exo* and *Exo* conditions (0.95 ± 0.005 and 0.91 ± 0.008, respectively) and the same behaviour in the RMSE ($F_{1,5} = 15.29$, $p = 0.001$), where we detect an increase in the tracking task error wearing the hybrid exoskeleton ($10.60 \pm 0.41°$) compared to the *No Exo* condition ($7.73 \pm 0.31°$). Looking at the delay between the reference trajectory and the subject's movement in the two conditions, we detect a significant increase in time delay only for the elbow ($p = 0.031$) when wearing the device ($0.735 \pm 0.051\,\mathrm{s}$) compared to the *No Exo* condition ($0.445 \pm 0.035\,\mathrm{s}$). The shoulder showed no significant differences in the execution of trajectories and the delay in *Exo* ($0.458 \pm 0.041\,\mathrm{s}$) and *No Exo* ($0.333 \pm 0.018\,\mathrm{s}$) conditions remained unaltered.

B. Muscular activity

In Figure 5a we report the EMG envelope of the two main muscles involved in the flexion/extension of the elbow. Figure 5a shows the EMG of a representative subject, averaged across trajectories and repetitions for each of the two conditions. In the elbow flexion corresponding to the "Rise" phase (Figure 5c left) we observe a significant reduction in the *biceps* activity ($p = 0.024$) while wearing the device ($1.92 \pm 0.12\%$) compared to the *No Exo* condition ($3.07 \pm 0.20\%$), with a reduction of $-32.53 \pm 3.52\%$ of the EMG activity. There are no changes in the *triceps* activity. On the other hand, during elbow extension corresponding to to the "Fall" phase (Figure 5c right), we find a reduction of the *biceps* activity ($-27.6 \pm 12.4\%$), but an increase of the *triceps* activity of $6.36 \pm 10.6\%$ which comes from the action that the wearer applies to counteract the gravity compensation from the exoskeleton and following the desired trajectories during the task. Figure 5b shows the EMG of a representative subject, averaged across trajectories and repetitions during the phases of shoulder abduction/adduction. During shoulder abduction (Figure 5d left for "Rise"), the hybrid exoskeleton significantly reduces the EMG activity of the *anterior* ($p = 0.016$) and *medial deltoids* ($p = 0.018$) by $-30.99 \pm 3.09\%$ and $-21.74 \pm 2.96\%$, respectively. Moreover, despite just marginally significant ($p = 0.0625$), it looks to provide an average change of $-28.91 \pm 2.48\%$ in the *posterior deltoid* also. In shoulder adduction ("Fall") (Figure 5d right) there are no significant changes in the EMG activity except for the *anterior deltoid* ($p = 0.024$) where we observe a change of $-32.60 \pm 2.42\%$ between the *No Exo* ($18.84 \pm 0.49\%$) and *Exo* ($7.06 \pm 0.22\%$) conditions. Looking at the CCI (Figure 6), we observe a slight reduction of the index in the main muscles involved in the elbow motion (i.e. *biceps* and *triceps*) between the condition *No Exo* ($43.58 \pm 2.30\%$) and *Exo* ($34.93 \pm 1.80\%$) but there is no statistical evidence of changes between the conditions. Likewise, taking into account the CCI of the main muscles involved in the shoulder motion, we have no evidence of statistical difference wearing the hybrid exoskeleton ($71.08 \pm 1.20\%$) compared to the *No Exo* condition ($75.28 \pm 1.12\%$).

IV. Discussion

World manufacturing companies are progressively considering the adoption of OEs [9]. The diffusion of such technologies could receive a further boost extending their assistive functionality to other joints, enhancing their capability to reduce muscle fatigue and related ergonomics risk. The integration of soft wearable devices to complement currently available OEs, without excessively increasing the overall system complexity or lowering their acceptability and usability, might be a viable solution to widen their application range. The hybrid solution,

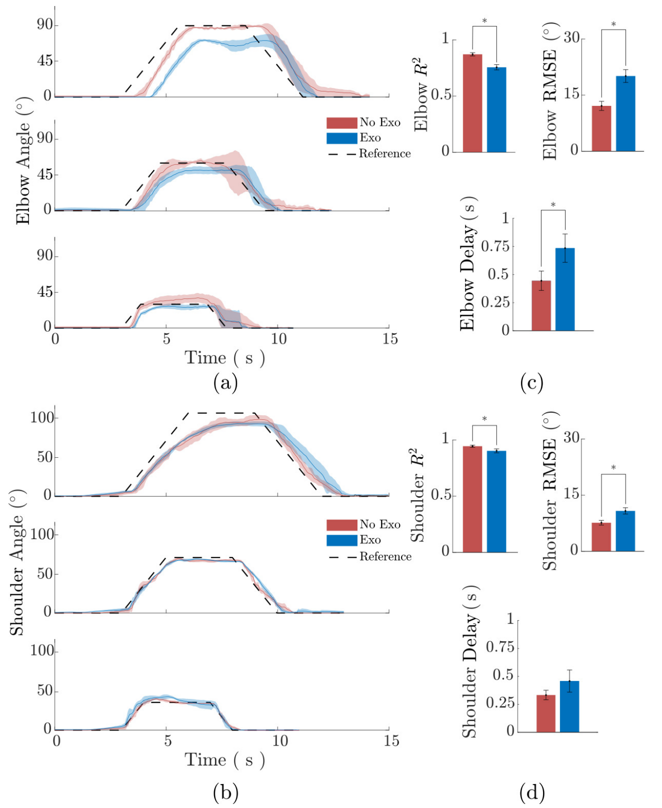

Fig. 4. *Tracking accuracy*: (a-b) time series of the trajectories followed by the elbow(a) and the shoulder (b) of a representative subject during the tracking task ($mean \pm SE$); (c-d) On the top the coefficient of determination r^2 and RMSE between the reference trajectory and the elbow (c) and shoulder (d) angle averaged across subjects, on the bottom the delay between the reference trajectory and the user's motion.

described and preliminary tested in our contribution, aims to merge two distinct approaches which have not yet intersect each other: our results showed how the exoskeleton affects the physiological motion and whether favorable or detrimental effects of its use could be observed through muscular activation patterns of the user.

A. Is it hindering the user?

Molding a wearable device around the user's body, efficiently providing assistance and assuring transparency to physiological movement are the key goals of wearable assistive technology. In the present work, we therefore assessed the subject's accuracy in following trajectories displayed by the virtual dummy via a visual interface to evaluate whether the hybrid exoskeleton fulfills the aforementioned requirements. As shown, wearing the device marginally decreases the accuracy of the movements, in particular at the elbow level, which was the joint assisted via active electromechanical support. It is reasonable to assume that the force tracking accuracy of the admittance controller, which tends to introduce a delay between the onset and the actual initiation of the movement, is responsible for the deviation from physiological trajectories. In a previous contribution, we observed the same detrimental effect which increases with the velocity of movement and was limited by the actuation bandwidth [23] and the static/Coulomb friction between the wire rope and the Bodwen sheath. In fact, different velocities of the movements affect the tracking accuracy proportionally to the speed of motion, as already explained exhaustively by a previous study on physiological and kinematic effect of an upper-limb exosuit employing a similar control strategy [25]. Variation of tracking accuracy is also observable at the shoulders, which were passively assisted, yet the tracking error was lower respect the one quantified for the elbows. A lower accuracy in trajectory tracking is expected considering that participants were naive to the device: further studies should include prolonged observations and analyze possible learning effects in the use of the technology. Keeping in mind the aforementioned digressions, the main purpose of the current contribution was to propose a novel architecture based on hybrid assistance and demonstrate that its use produces, on short term, no significant effects on physiological movement. The analysis of the kinematics reports significant difference in the delay of the subjects' movements in the *Exo* condition during elbow flexion/extension, a behaviour less noticeable for the shoulder elevation. However, taking into account the diverse approaches to assist the distal and proximal joints such difference seems not to dramatically alter the inter-joint coordination. We may also speculate that a prolonged use of the device and additional training for the wearers might play a role in better performance and adaptation to the different natures of the hybrid assistance.

B. Is it changing the muscular activity?

Considering that one of the main purposes of industrial exoskeletons is reducing metabolic expenditure, we focused our investigation on extracting such information by observing the activity on the main muscles involved in the tasks. The changes in the muscle activation levels on the main muscles acting on the shoulder joint (anterior, medial and posterior parts of the deltoid and upper trapezius) are comparable with the one reported using the passive part of the hybrid exoskeleton as a stand alone device [9]. The hybrid exoskeleton, with respect to the stand alone original device, successfully reduces muscular activation on the shoulders as well as the elbows: a reduction of 32% of *biceps* activation during elbow flexion has been reported and provided by the actuated layer, but of course we also have to consider that employing actuation and force control may require wearers to use other muscles to drive the exosuit in the directions which are not actively supported. This is confirmed by our data extracted from the muscular activity of the *triceps* which showed an increase of 6% in the extension phase with respect to the *No Exo* condition. A similar results has been previously found in other investigations [18]: to such an extent, we decided to run our analysis to check whether co-contraction and shifting of muscular activation were still significantly hampering the efficacy of the device. We found that the effort required for co-contraction and additional muscular activities, requested to drive the hybrid device, have been outweighed by the net decrease of muscular effort in the assisted joints, and this can be considered as a positive effect of the proposed technological solution. While EMG model-based control approaches can avoid this problem [26], we

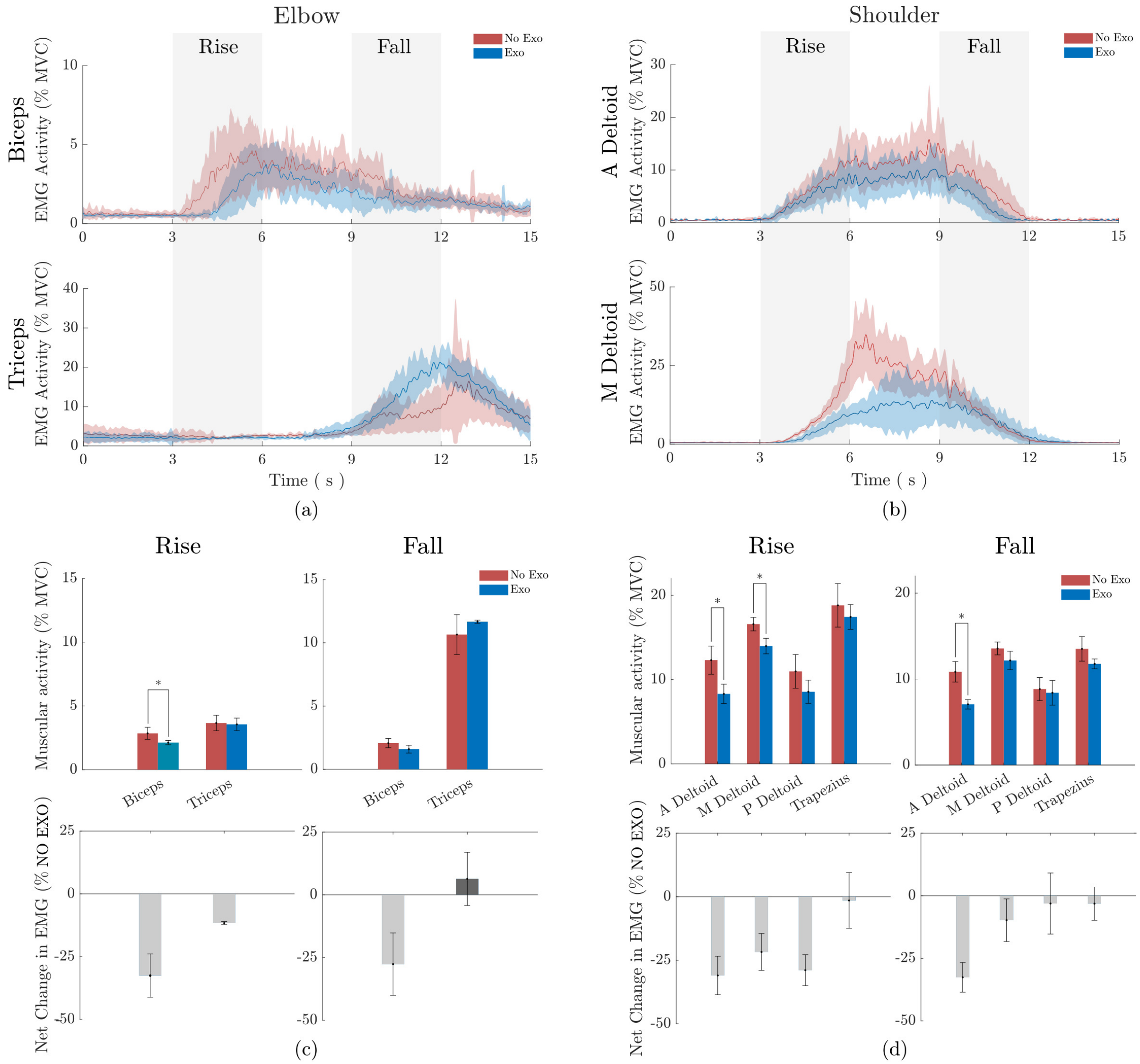

Fig. 5. *Muscular activity*: (a) time series of the *biceps* and *triceps* activity of a representative subject, averaged across the repetitions, with focus on the phases of elbow flexion (rise) and extension (fall); (b) time series of the muscles of a representative subject, averaged across the repetitions, presenting significant changes in the activity on the phases of shoulder abduction (rise) and adduction (fall); (c) EMG activity and net change between the conditions *No Exo* and *Exo* of the main muscles involved in the elbow movements averaged across subjects in the phases of flexion (left) and extension (right); (d) EMG activity and net change between the conditions *No Exo* and *Exo* of the main muscles involved in the shoulder movements averaged across subjects in the phases of abduction (left) and adduction (right).

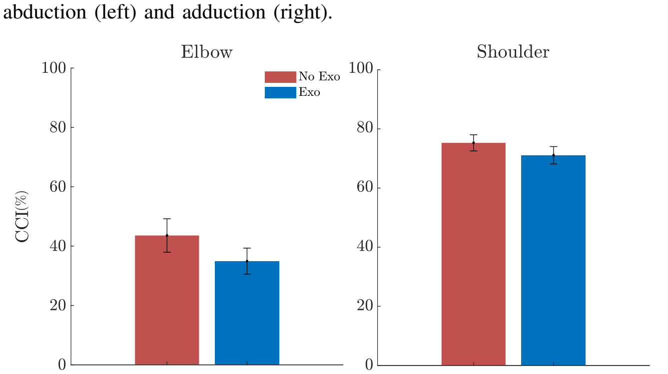

Fig. 6. *Co-Contraction Index*: On the left panel CCI between *biceps* and *triceps* in the elbow motio while on the right panel CCI calculated on the *deltoids* in the shoulder movements

believe that in a daily use case such a solution would only add further complexity to the system which instead should be characterized by a portable, calibration-free, still intuitive architecture. Notwithstanding, the limitations of the study are multiple: the first is the controlled environment which does not reflect the variability of areal industrial scenario and does not contemplate all the variety of actions to which a worker is exposed. The second, and equally important limitation, is the possible impact deriving from the increase of weight of the device on the lumbar region. Indeed, concerning this prototype, the overall weight increased from $4.2\,\mathrm{kg}$ of the commercial exoskeleton to $6.4\,\mathrm{kg}$ because of the integration of the exosuit. The third limitation of the current study regards the number of acquired EMG channels to understand how the device affects users' muscular activation: to provide a more accurate evaluation, future investigations will employ a wider number of muscle groups within more complex tasks involving multi-joint coordination. In the coming years, we expect that research activities should be carried out along two main directions. Firstly, an engineering effort should be spent on the reduction of the overall weight. We can expect that a future, lighter version of the adopted commercial exoskeletons could be a suitable starting point (e.g., recently COMAU has presented the XT version with a weight of $3\,\mathrm{kg}$ [https://mate.comau.com/]). Then, it is possible to envision that the design of the mechatronic components of the exosuit (control electronics and transmission) could be revised to minimize the weight. Secondly, assuming that a reasonable final weight could be in the range $4.5 - 5$ kg, it will be paramount carrying out investigations to assess the physiological impact of prolonged use of the hybrid device on the user's spine, posture and balance. Clearly, these kinds of challenges are multifaceted and require the involvement of multidisciplinary effort, including engineers, clinicians, psychologists, healthcare managers and industrial representatives [27].

V. Conclusions

Within our study, we had the chance to demonstrate that two approaches, which usually define two separate domains, can be integrated to provide complementary actions: we have demonstrated that active and passive assistance, if opportunely combined, can efficiently work to provide tangible improvement in terms of muscular efficiency. In order to envision a large consumer technology, we designed a dual-layer system, utilizing soft and rigid components, that is an intuitive and ready-to-use solution. However, wearable industrial technologies must be rigorously tested at the system level, by accurate observations of real use and, above all, embracing disciplines like neuroergonomics and psycho-physics which have not been contemplated in our initial study. Only by a multidisciplinary approach, we will be able to determine if such a technology will be effective in preventing work-related-musculo-skeletal disorders and will be accepted by a wide audience as everyday use protective gear.

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