

Article

Design and Evaluation of a Novel Passive Shoulder Exoskeleton Based on a Variable Stiffness Mechanism Torque Generator for Industrial Applications [†]

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Abstract: Work-related musculoskeletal disorders (WMSDs) are a common occupational health problem in industries, and they can lead to decreased productivity and a reduced quality of life for workers. Exoskeletons, as an emerging technology, have the potential to solve this challenge by assisting arm movements and reducing muscle effort during load lifting tasks. In this paper, a passive exoskeleton based on a variable stiffness mechanism (VSM) torque generator is proposed and evaluated. This exoskeleton can provide adjustable torque curves and accommodate three degrees of freedom (DOFs) while remaining compact and lightweight. The workspace analysis shows that the workspace of this exoskeleton is sufficient for most industrial manual handling tasks. The experimental results demonstrate that the exoskeleton effectively reduces muscle effort during overhead reaching and load-lifting tasks, highlighting its effectiveness for repetitive tasks in industrial settings.

Keywords: industrial exoskeleton; variable stiffness mechanism; torque generator; muscle activity evaluation; work-related musculoskeletal disorders



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1. Introduction

Exoskeletons are wearable machines worn on the human body for motion assistance and power augmentation in the form of either upper-body, lower-body, or full-body exoskeletons [1,2]. They have been found in many applications, ranging from medical rehabilitation to industrial settings [3], one of which is to prevent WMSDs for physical workers in various industries. WMSDs include disorders in the muscle, skeleton, and related tissues. This is a common occupational health problem in various industries [4]. Approximately half of WMSDs affect the shoulder and lower back muscles of industrial workers who perform repetitive tasks at overhead heights [5]. WMSDs are particularly prevalent in the manufacturing industry [6–8], where some manual handling tasks are difficult to robotize, making human workers essential and central [9]. WMSDs consequently affect worker productivity, incur healthcare costs, and impact quality of life due to limited mobility and increased rehabilitation costs [10,11]. In Germany, for example, muscle disorders caused a total loss for 47.6 billion in 2016, which included both the social expenses and loss of labor productivity [12]. Thus, industries are showing a growing interest in exoskeletons to aid workers in manual handling tasks and mitigate the risk of WMSDs, benefiting both employers and employees.

Exoskeletons, by assisting human limb movements, can effectively reduce muscle effort. Based on the energy source, exoskeletons can be classified into active exoskeletons and

passive exoskeletons. Active exoskeletons include actuation systems that generate torque actively, such as electrical [13–15], pneumatic [16–18], and hydraulic actuations [19–22]. However, active exoskeletons are, in general, complicated, and they require on-board batteries and control units, making the entire system bulky and heavy [23]. Furthermore, the control system must be properly designed to provide the needed torque and ensure stability [24,25]. Rather than adopting active actuators, passive exoskeletons can provide assistive torque through the stored potential energy in elastic elements, which are in the form of either mechanical springs [26–29] or gas springs [30–32]. Notably, researchers also utilize magnetic springs on the passive exoskeletons to compensate for the gravity acting on the upper limbs [33,34]. Compared to active exoskeletons, passive exoskeletons are typically lighter and less complex, since there is no actuator or external power source needed [35]. In industrial applications, passive shoulder exoskeletons enable workers to slightly enhance their load-lifting capabilities while significantly reducing muscle fatigue during repetitive manual handling tasks [36]. Examples of commercial passive shoulder exoskeletons include Skelex 360 [37], Vex [38], and Paexo [39], adopting either leaf springs or linear springs, and EVO [40] which adopts a gas spring can provide 2.2–6.8 kg of lifting force to each arm. For these exoskeletons, overcompensation, which describes the condition where the exoskeleton provides more assistance than required, can reduce the muscle activities involved in shoulder flexion. However, in the case of extension, muscle activities are increased, which means that the exoskeleton impedes the natural movement of the subjects [41]. Therefore, it is desirable to make the torque profile adjustable for different individuals to avoid overcompensation. In addition, exoskeletons remain bulky and heavy, which impedes their widespread applications in the industry. Therefore, more compact and lightweight passive exoskeletons are needed for high-performance assistance and also for their widespread application.

In this paper, a passive shoulder exoskeleton that incorporates the variable stiffness mechanism (VSM) principle is presented. By adopting the VSM, a novel torque generator that is both compact and capable of providing adjustable assistance was developed. This work is based on our previous research [42] in which we presented the foundational principles of the shoulder mechanism design. In this study, a portable shoulder exoskeleton that can accommodate the three primary motions of human shoulder joints was developed. To assess the effectiveness and practicality of this exoskeleton in real-world scenarios, a series of load lifting experiments were conducted. These experiments were designed to evaluate the exoskeleton's performance in reducing muscle strain and improving overall efficiency by mimicking manual handling tasks.

This paper is organized as follows: In Section 2, the concept of the torque generator with the variable stiffness mechanism (VSM) principle is introduced. Section 3 describes the mechanical design of the passive shoulder exoskeleton. Section 4 presents a comprehensive workspace analysis of the exoskeleton, evaluating its range of motion and ability to accommodate the three primary motions of human shoulder joints. Section 5 describes the preliminary load lifting experiments carried out to assess the exoskeleton's performance. This work is discussed and concluded in Section 6.

2. Basic Design Concept of the VSM Torque Generator

The VSM was first proposed by Li and Bai in [43], then applied to the design of exoskeletons [44]. Here, we provide a brief overview of the VSM principle for completeness.

The design of the VSM is based on a four-bar linkage. Figure 1a illustrates a specific case where bar 4 becomes zero-length and bar 2 is replaced with an elastic element. Building upon this zero-length base link four-bar linkage concept, a new mechanism is implemented using a pulley cable system connected to a linear spring, as shown in Figure 1b. Here, bar 1 and bar 3 can be seen as the input and output levers, respectively.

Figure 1c demonstrates a torque generator concept implemented with input and output plates coupled with pulleys and cables to achieve varying stiffnesses. The configuration shown in Figure 1b is replicated to expand the stiffness range.

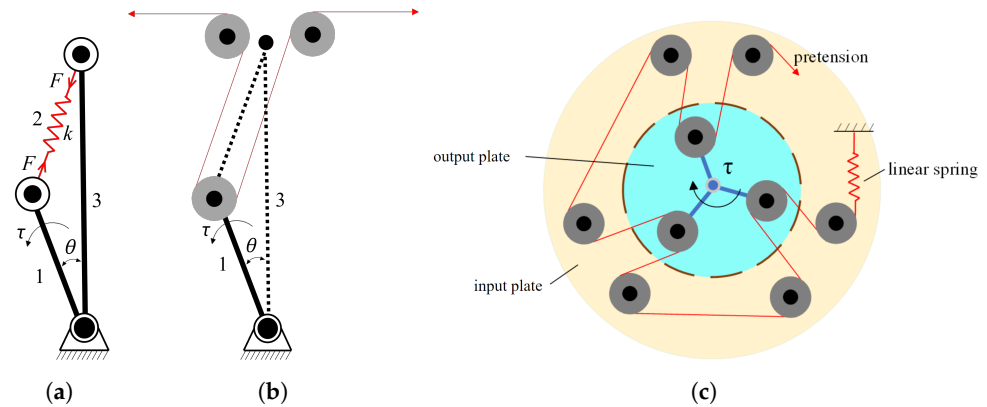


Figure 1. The basic concept of the VSM. (a) The special case of the four-bar linkage, (b) the implementation of the VSM with pulleys and cable, (c) torque generator concept with VSM, where the torque is generated due to the restoring force of the spring.

As shown in Figure 1a, the force applied to link 1 equals the force generated by the spring. Therefore, the torque applied to link 1 can be expressed as follows:

$$\tau = J \cdot F \quad (1)$$

where F is the tension force in the spring and J is the Jacobian, which can be further written as follows:

$$F = k \cdot \Delta l_2 + F_0 \quad (2)$$

$$J = \frac{l_1 l_3 \sin \theta}{\sqrt{l_1^2 + l_3^2 - 2l_1 l_3 \cos \theta}} \quad (3)$$

where l_1 and l_3 are the lengths of bar 1 and bar 3 shown in Figure 1a, respectively. θ is the deflection angle between bar 1 and bar 3. k , Δl_2 , and F_0 are the stiffness, elongation, and pretension force of the spring, respectively.

Figure 1c shows the torque generator utilizing VSM, constructed with cables and multiple pulleys. In this design, the output torque is expressed as follows:

$$T_{VSM} = c^2 \cdot N^2 \cdot J \cdot k \cdot \Delta l_2 + c \cdot N \cdot J \cdot F_0 \quad (4)$$

Here, the geometrical factor c denotes the influence of mechanical geometry on cable elongation, while N represents the number of pulleys on the output plate.

Figure 2 illustrates the torque distribution of the VSM concerning the deflection angle and pretension, computed using Equation (4), with the parameters set to $c = 2$, $N = 1$, $k = 5.4 \text{ N/mm}$, $l_1 = 22.4 \text{ mm}$, and $l_2 = 32.5 \text{ mm}$.

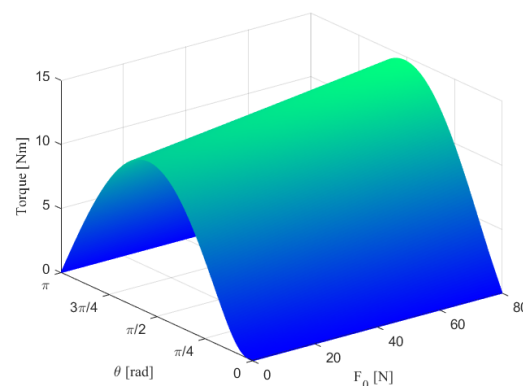


Figure 2. Torque distribution of the VSM with respect to pretension force and deflection angle.

3. Design Implementation of the Exoskeleton

3.1. Parameter Optimization of the VSM

As shown in Figure 2, the torque generated by the VSM depends on various parameters, including geometric parameters, material properties, and the layout of the pulley–cable–spring system.

Considering the range of rotation of the shoulder joint, only one pulley is utilized to avoid interference between the moving pulley and the central axis and to maximize the rotation of the joint. Moreover, the required torque should closely match or equal the torque needed to lift the arm. This can be achieved through design optimization.

Utilizing the lengths and masses of the arm segments, a 7-degrees-of-freedom (DOF) model of the human arm was first developed, similar to the approach outlined in [27]. With this model, a resulting torque curve for motion in the sagittal plane can be calculated. An objective function was then defined, aiming to minimize the difference between the required torque of the human arm and the torque provided by the exoskeleton.

The optimal design was implemented using an interior point algorithm [45]. The objective function in the optimization is given as follows:

$$f(x) = \left(\int_{\beta_l}^{\beta_u} \int_{\alpha_l}^{\alpha_u} [w \cdot (T_S(\beta, \alpha) + T_{VSM}(\beta, x))]^2 d\alpha d\beta \right)^{1/2} \quad (5)$$

where $x = [l_1, l_3, k, F_0]^T$, which contains the parameters of the VSM that need to be optimized. Here, T_S is the shoulder torque, which can be found in [42]; T_{VSM} is the VSM torque, calculated using Equation (4); and w is a weight function used to amplify certain positions over others. The boundaries (α_l, α_u) and (β_l, β_u) are determined by the range of joint rotations, which are $\alpha_l = \beta_l = 0^\circ$, $\alpha_u = 150^\circ$ and $\beta_u = 180^\circ$. α and β are the elbow and shoulder angles, respectively, as illustrated in Figure 3.

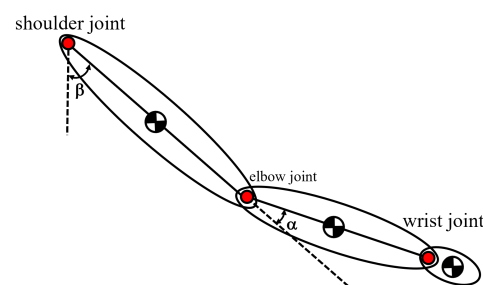


Figure 3. Definition of the shoulder and elbow extension/flexion angles with a simplified model of the human arm.

The weight function is built with an emphasis on the position of overhead work, with its maximum at $\beta = 110^\circ$ and $\alpha = 47^\circ$. The optimization problem confines the design space to geometrically and logically feasible intervals, as outlined in [42]. Considering the design requirement to support 100% of the arm weight for an adult at the 50% percentile of Danish adults [46], optimal parameters were determined, as presented in Table 1. Figure 4 displays the torque curve corresponding to this configuration of the torque generator.

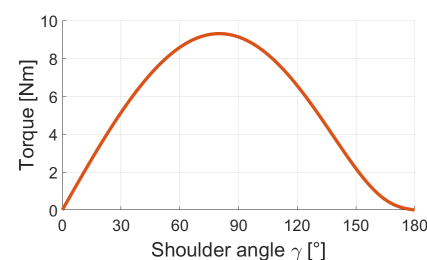


Figure 4. Torque curve with optimized parameters.

Table 1. Torque generator parameters.

l_1	l_3	k	F_0
17.4 mm	31.4 mm	6.2 N/mm	10 N

3.2. Prototype Development

Using the optimal parameters obtained, a prototype of the shoulder exoskeleton was developed. Figure 5a–c show the CAD model of the exoskeleton. An encoder was included in the design for angle recording.

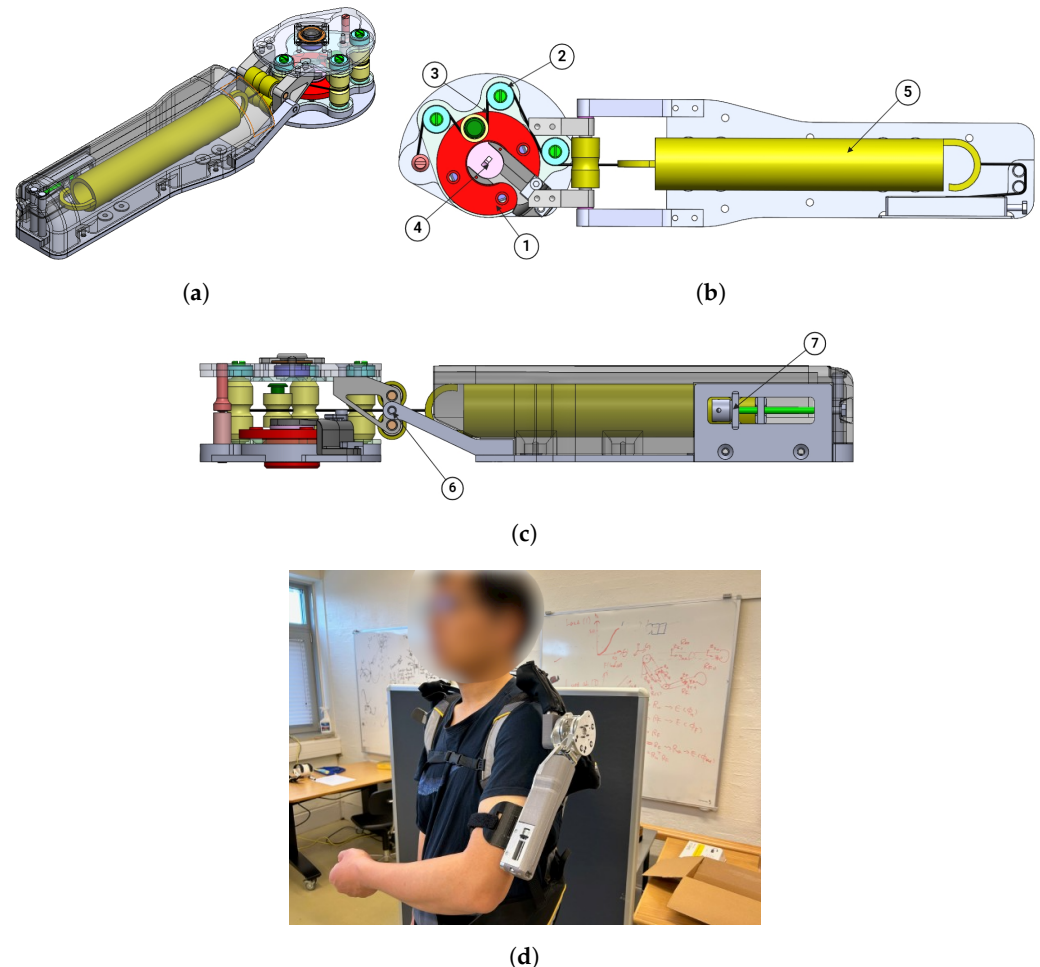


Figure 5. Views of the shoulder exoskeleton. (a) Isometric view of the torque generator, (b) top view of the torque generator with the cover removed: (1) input plate, (2) output plate, (3) cable, (4) encoder, (5) linear spring, (c) side view with (6) free revolute joint, (7) manual adjusting module. (d) The shoulder exoskeleton worn by the lead author.

Furthermore, to be adaptive to different users and assistance requirements, a manual adjusting module has been designed to change the pretension force of the spring, enabling adjustable assistance levels. The prototype, mounted on a frame manufactured by Skelex Inc. and worn by the lead author, is shown in Figure 5d.

Table 2 outlines the key specifications of the exoskeleton. In addition to the shoulder movement in sagittal plane, the exoskeleton allows shoulder movement in frontal and transverse planes, thus facilitating shoulder motions in all three DOFs.

Table 2. Main specifications of the shoulder exoskeleton.

Description	Value
Length (without frame)	58.6 mm
Width (without frame)	293.8 mm
Weight (without frame)	0.705 kg
Number of DOFs	3
Range of extension/flexion	$-36^{\circ} \sim 178^{\circ}$
Range of abduction/adduction	$-5^{\circ} \sim 45^{\circ}$
Range of internal/external rotations	$-5^{\circ} \sim 81^{\circ}$
Designed payload	0~5 kg
Spring model	BUFSP22-2.5-91

4. Workspace Analysis of the Exoskeleton

For the developed exoskeleton, the workspace of the exoskeleton is analyzed to ensure the necessary range of motion for various industrial tasks. Denavit–Hartenberg (DH) is adopted for this purpose. To simplify the calculation, the shoulder joint is modeled as a spherical joint, and the elbow pronation/supination are ignored. The DH parameters are summarized in Table 3.

Table 3. DH parameters.

Link (i)	θ_i	d_i	a_i	α_i
1	θ_1	0	0	$\frac{\pi}{2}$
2	θ_2	0	0	$\frac{\pi}{2}$
3	θ_3	L_u	0	$-\frac{\pi}{2}$
4	$\frac{\pi}{2} - \theta_4$	0	L_f	0

Here, L_u and L_f represent the length of the upper arm and forearm, respectively. The ranges of θ_1 , θ_2 , and θ_3 are the angles of extension/flexion, abduction/adduction, and internal/external rotation, respectively, as shown in Table 2. θ_4 is in the range of $(0, 153^{\circ})$, which is the angle of elbow flexion/extension. The transformation matrices are readily obtained using the DH parameters:

$$A_i = \begin{bmatrix} \cos(\theta_i) & -\sin(\theta_i) \cos(\alpha_i) & \sin(\theta_i) \sin(\alpha_i) & a_i \cos(\theta_i) \\ \sin(\theta_i) & \cos(\theta_i) \cos(\alpha_i) & -\cos(\theta_i) \sin(\alpha_i) & a_i \sin(\theta_i) \\ 0 & \sin(\alpha_i) & \cos(\alpha_i) & d_i \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (6)$$

The overall transformation matrix from the base frame (human trunk) to the end-effector (human hand) frame can be expressed as follows:

$$T = A_1 \cdot A_2 \cdot A_3 \cdot A_4 = \begin{bmatrix} R & P \\ 0 & 1 \end{bmatrix} \quad (7)$$

where P is the reaching position of the human hand. The Monte Carlo sampling method is used to generate pseudo-random numbers between 0 and 1, adhering to a uniform distribution. Each sampling produces a set of values for each joint [47]. The arm motion values generated by sampling pseudo-random numbers can be expressed as follows:

$$\theta_i^j = \theta_i^{\min} + (\theta_i^{\max} - \theta_i^{\min}) \cdot \text{rand}(N, 1) \quad (8)$$

where $\text{rand}(N, 1)$ represents the pseudo-random number in the range of $(0, 1)$ complying with the uniform distribution. N is the sampling time, which is set to 3000 in this simulation. Figure 6 shows the workspace's point cloud of the human arm with the exoskeleton. Compared with the results of the workspace analysis for the human arm in [48], it can be

concluded that the workspace of the arm is not restricted by the shoulder exoskeleton for general industrial tasks.

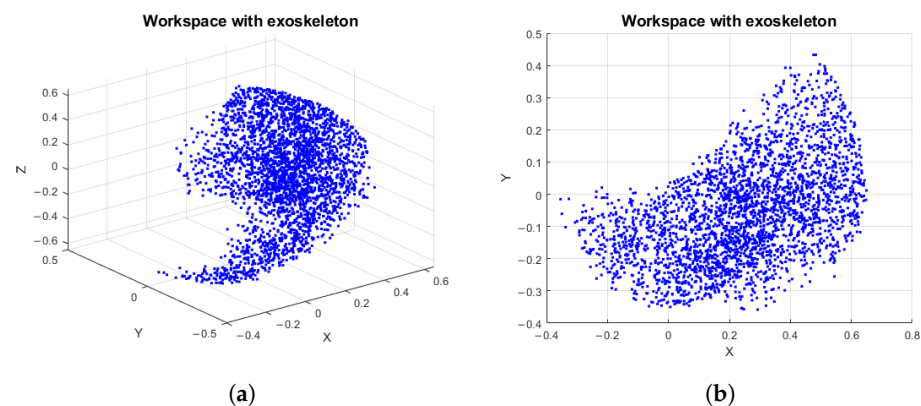


Figure 6. Workspace point cloud of the human arm with the exoskeleton. (a) Isometric workspace, (b) workspace projected into the XOY plane. The units in these plots are meters.

5. Performance Evaluation of the Exoskeleton

5.1. Experimental Protocol

Experiments were conducted on the prototype at the Exoskeleton Lab, Aalborg University, Denmark. This study involved three right-handed male participants aged 25–30, none with a recent history of musculoskeletal disorders. The participants provided written informed consent before the study. Prior to the experimental trials, the participants engaged in a ten-minute training session to familiarize themselves with the exoskeleton and load lifting tasks. During this session, adjustments were made to the exoskeleton's mounting interface to ensure maximum comfort.

Electromyographic (EMG) signals were utilized to assess the exoskeleton's performance. EMG data were captured using the FREEEMG system (BTS S.p.A, MI, Italy). This study focused on the activity of four shoulder-related muscles: the upper trapezius (TR), biceps brachii (BB), and the anterior and middle deltoids (AD and MD). Electrode placement followed the guidelines recommended by Cram [49]. Figure 7 illustrates the placement of the electrodes.



Figure 7. Placement of the electrodes.

To evaluate the exoskeleton's performance, both static and dynamic load lifting experiments were conducted. The maximum voluntary contractions (MVCs) of the aforementioned muscles were recorded for each participant prior to the experiments.

In the static load lifting experiment depicted in Figure 8, the participants were instructed to hold a 2 kg weight at 90 and 135 degrees, both with and without the exoskeleton, for 10 s in each of five trials. To prevent muscle fatigue, the participants rested for 3 min between trials.

In the dynamic load lifting experiment shown in Figure 9, the participants lifted a 2 kg weight from 0 to 135 degrees, both with and without the exoskeleton, for five trials. Upon reaching 135 degrees, they maintained the position for 3 s. Programmed voice

guidance was used to instruct the participants while performing these tasks. To prevent muscle fatigue, the participants rested for 3 min between trials.

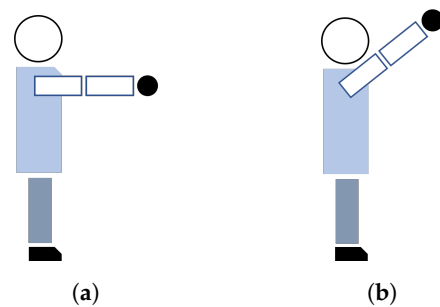


Figure 8. Illustration of static load lifting experiments. (a) Load lifting at 90 degrees, (b) load lifting at 135 degrees.

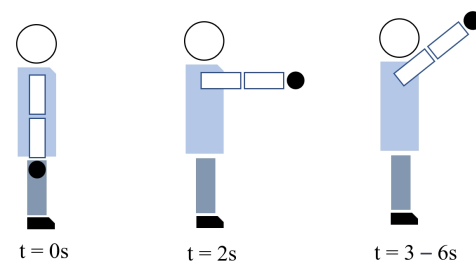


Figure 9. Illustration of dynamic load lifting experiment where participants were asked to lift the load from 0 to 135 degrees and hold it at 135 degrees for 3 s.

5.2. Results

The EMG signals were sampled at a rate of 1000 Hz. Initially, the raw EMG signals were filtered using a third-order bandpass Butterworth filter with a passband of 20–450 Hz, followed by full-wave rectification. Subsequently, the signals were processed using a lowpass filter with a cutoff frequency of 10 Hz. Figure 10 presents an example of both the raw and processed EMG signals from one trial of the static load lifting experiment.

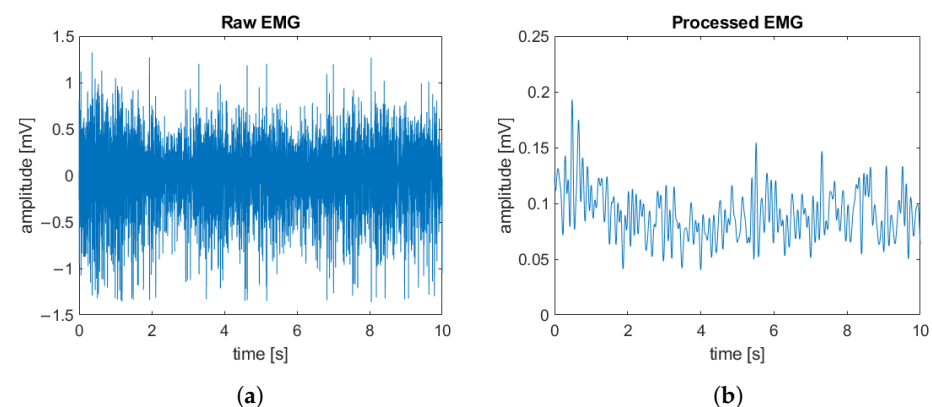


Figure 10. Raw and processed EMG signals recorded during the static load lifting experiment. (a) Raw EMG signal, (b) processed EMG signal.

All the EMG signals were then normalized to the MVC. The mean normalized muscle activation for the relevant muscles was extracted for both the static and dynamic load lifting tasks.

Figures 11 and 12 display the normalized muscle activation during the static and dynamic load lifting experiments for the three subjects, respectively. The average reductions in muscle activation with and without the exoskeleton across the different tasks are summarized in Table 4. It is evident that the muscle activation in the anterior deltoids and

biceps brachii was significantly reduced across all the tasks, particularly for the overhead lifting tasks, such as load lifting at 135 degrees, both in the static and dynamic conditions.

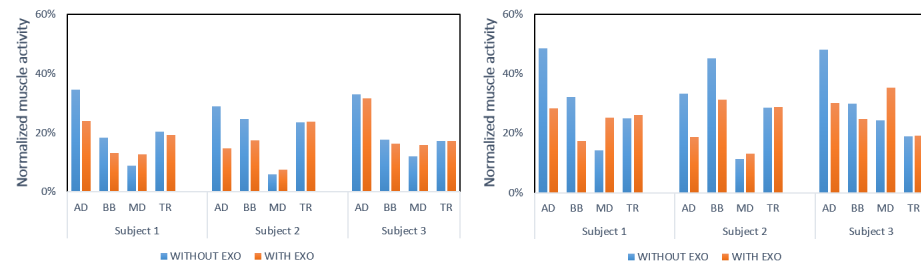


Figure 11. Normalized mean muscle activation during static load lifting of 90 degrees (left) and 135 degrees (right).

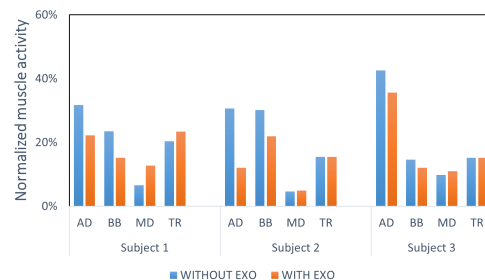


Figure 12. Normalized mean muscle activation during dynamic load lifting.

Table 4. Averaged reduction in muscle activation during load lifting tasks.

Tasks	Muscle Activation Reduction			
	AD	BB	MD	TR
Static load lifting (90°)	8.72%	4.55%	−3.16%	0.34%
Static load lifting (135°)	17.58%	11.37%	−7.82%	−3.62%
Dynamic load lifting	11.68%	6.36%	−2.57%	−1.01%

In addition to the anterior deltoids and biceps brachii, which are involved in shoulder flexion, the muscle activities of the upper part of the trapezius and middle deltoids were also assessed to evaluate the potential side effects of the exoskeleton. The results indicated that the exoskeleton had no significant impact on the upper trapezius, which shows that the exoskeleton did not restrict the elevation of the scapula.

In our recent work [50], the Skelex 360 exoskeleton was evaluated using a series of tasks in a controlled laboratory environment. Table 5 shows the muscle activation reduction results in the box moving task. The posterior deltoids (PDs) are the additional muscles involved in the experiment. Although one muscle involved in the evaluation differed, the overall methodology and conditions for the box moving task are comparable.

Table 5. Average reduction in muscle activation during box moving with Skelex 360.

Task	Muscle Activation Reduction			
	AD	BB	MD	PD
Box moving	7.65%	−1.60%	1.73%	−2.02%

Compared to the Skelex 360 exoskeleton, our proposed exoskeleton has a larger muscle activity reduction in the anterior deltoids and biceps brachii. However, different from Skelex 360, all of the subjects experienced impedance in the middle deltoids with our proposed exoskeleton, likely due to the exoskeleton's weight. Furthermore, misalignment

between the human body and the exoskeleton may restrict free movement. Future improvements could involve integrating quasi-passive actuation to mitigate impedance [51], for which human impedance detection and human–robot interaction modulations can be included [52]. Additionally, employing a unique parallel structure could potentially eliminate misalignment concerns [53].

6. Discussion and Conclusions

In this work, we designed and evaluated a portable passive shoulder exoskeleton based on a variable stiffness mechanism to assist workers in industrial applications. The test results indicated that the exoskeleton effectively reduced the muscle activities in the anterior deltoids and biceps brachii, particularly during overhead tasks such as static load lifting at 135° . The reduction in muscle activation for load lifting position at 90° was decreased. This suggests its potential as a valuable tool for alleviating muscle strain among workers. However, there was an increase in the muscle activities in the middle deltoids. To address this, optimization strategies such as reducing weight through topology optimization and minimizing misalignment through ergonomic design could be implemented.

Additionally, considering the weight function's emphasis on the position at $\beta = 110^\circ$ and $\alpha = 47^\circ$, further improvements can be made by adding an adjustment module to customize optimal working positions. This would enhance the exoskeleton's adaptability to a wider range of tasks.

A workspace analysis revealed that the exoskeleton provided a sufficient workspace for general manual handling tasks, making it suitable for industrial applications. However, when no payload is present, the subjects may need to exert effort to overcome the torque provided by the exoskeleton. Introducing a clutch mechanism to disengage assistance when walking without a payload could enhance usability.

Future research could focus on assessing the feasibility of implementing the exoskeleton in real-world work environments and further optimizing its design for improved comfort and adaptability.

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