Back-Support Exoskeleton Control Using User's Torso Acceleration and Velocity to Assist Manual Material Handling

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Back-Support Exoskeleton Control Using User's Torso Acceleration and Velocity to Assist Manual Material Handling

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Abstract This work analyzes the use of users' dynamics to define the assistance of a back-support exoskeleton for assisting manual material handling. Exploiting the acceleration and velocity of the user's torso on the sagittal plane allows to distinguish between lifting and lowering phases and accordingly adapt the assistance. Theoretical and practical issues of strategy implementation are discussed.

1 Introduction

In many industrial sectors, workers perform manual material handling (MMH) activities, that overload and compress the spine causing injuries and musculoskeletal disorders (MSDs). Back-support exoskeletons are being introduced in industries, where full-automation is not feasible and human's flexibility is required [1]. Exoskeletons promise to reduce the MSDs risk, by reducing the compression of the spine [2].

Our group has developed a torque-controlled back-support exoskeleton for which we explore different possible control strategies to suit the need for assistance. For improving the effectiveness of the assistance provided in dynamic MMH tasks with respect to state-of-the-art methods, valuable information can be obtained from the user's dynamics during the execution of the task. In a previous work [3] we presented a new strategy that uses the user's torso angular acceleration for assisting symmetric lifting and lowering, while a similar device [4] employed the torso angular velocity for the same purpose.

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This work analyzes the difference of designing a control strategy using the acceleration against using the velocity for assisting lifting and lowering tasks. The strategies were implemented on the XoTrunk back-support exoskeleton, an evolution of the prototype employed in [3].

2 Control Strategies

2.1 Rationale for Using User's Dynamics

Most current strategies for assisting lifting and lowering tasks act to compensate for the effect of gravity [1]. The assistance is based on the static characteristics of the user's movement (mainly torso inclination), and thus would not adapt to different task phases. An inclination-based strategy generates the peak assistance corresponding with the maximum torso flexion, as happens for a passive exoskeleton.

However, as clarified in [5], the lumbar moment reaches its peak at the beginning of lifting, i.e. when the user grasps the box and starts to lift it (after the maximum flexion occurred), because upper body mass and the mass of the load requires acceleration upwards. As a result, (1) an inclination-based strategy would not generate the peak assistance corresponding with the peak in torque need [5]. Moreover, providing the same assistive torque during the descent phase (lowering) and the ascent phase (lifting) (2) limits the maximum physical assistance, because increasing assistance in the latter corresponds to increasing hindrance in the former. Therefore, for inclination-based strategies, as for passive exoskeletons, the assistance provided for supporting the user during lifting must be scaled according to his acceptance of hindrance during lowering. because he has to accelerate upward his mass plus the mass of the box.

Measuring the dynamics of the user's torso allows to distinguish the lowering and the lifting phases, and thus assist them differently. If the angular acceleration of the user's torso is used to proportionally define the assistive torque, additional support is provided accordingly with the peak in the assistance need (1) i.e., when the user grasps the box and starts the lifting phase, accelerating his and the box's masses. On the other end, the reduction of the assistance due to the inclination-based torque during the lowering phase, achieved by employing the torso acceleration or velocity, reduces the hindrance for the flexion of the torso (2).

2.2 Implementation

Focusing on overcoming these two limitations (1) and (2), in a previous work [3] we presented a new strategy making use of the user's torso angular acceleration.

Additionally, we implemented a strategy based on user's torso angular velocity on the same exoskeleton for comparing the two strategies.

The angular velocity $\dot{\theta}_h$ is measured with an Xsens MTw IMU (Xsens Technology), attached to the user's torso (approximately at the sternum). The angular acceleration $\ddot{\theta}_h$ is obtained by differentiating and filtering the angular velocity on the sagittal plane (low-pass filter, cut-off frequency 1 Hz). Then, dynamic torques $\tau_{dynamic}$ are defined proportional to the torso angular velocity or acceleration, while the static torque τ_{static} is proportional to torso inclination. The assistive torque $\tau_{assistive}$ is finally computed as the sum of the static (inclination-based) and the dynamic (velocity or acceleration-based) torques, that can be scaled adjusting the respective control gains K:

$$\tau_{assistive} = \tau_{static} + \tau_{dynamic}$$

$$\tau_{static} = K_{incl}\ sin(\theta_h) \quad \text{inclination-based}$$

$$\tau_{dynamic} = \begin{cases} -K_{vel}\ \dot{\theta}_h & \text{velocity-based} \\ -K_{acc}\ \ddot{\theta}_h & \text{acceleration-based} \end{cases}$$

In the following, *acceleration* and *velocity* strategies indicates that the assistive torque is computed as the sum of the inclination-based with velocity or acceleration-based torques, respectively. The *inclination* strategy has $\tau_{dynamic} = 0$.

3 Experimental Results

Experiments were done with 9 subjects. The differences in lumbar compression peaks (estimated as in [2]) of *velocity* and *acceleration* strategy w.r.t the condition without the exoskeleton were statistically significant (ANOVA test, $p < 0.05$). The difference between *velocity* and *acceleration* strategy (compression force reductions w.r.t no exoskeleton equal to 11% and 15%, respectively) was not significant.

4 Discussion

A noticeable difference between velocity and acceleration signals emerges at the beginning of lifting (c), when the subject is accelerating upwards. As the acceleration increases, the *acceleration* strategy provides greater assistance which is in time

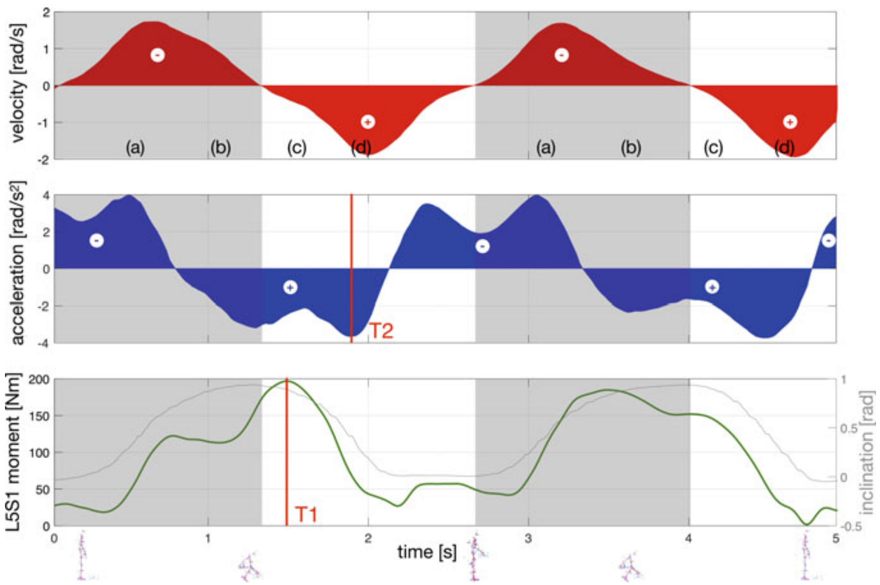

Fig. 1 The curves of a subject's torso angular velocity and acceleration are shown during box lifting and lowering. Their contribution to $\tau_{assistive}$ is indicated as: (+) the total torque is increased by $\tau_{dynamic}$, (−) if it is decreased. The lumbar moment generated at L5S1 disc (as estimated in [2]) is displayed at the bottom (green line). T1 is the instant of peak L5S1 moment, T2 is the instant of peak angular acceleration. To highlight task phases, trunk inclination and the human model position (at the bottom) are included; the phases are: beginning (a) and end (b) of lowering (grey background), beginning (c) and end (d) of lifting (white background)

with the subject need, as the lumbar moment (and hence the compression) reaches its maximum value (1). Conversely, the velocity starts to increase at the beginning of lifting and reaches its maximum after the lumbar moment peak, actually increasing the assistance at the end of lifting (d). Results show larger compression force reductions for the *acceleration* strategy, although not significant. Further analysis is needed to reveal the benefits of the two strategies, focusing on other aspects of the assistance, e.g., subjective perception of support and hindrance or changes in the execution speed.

With respect to an *inclination* strategy, another advantage of using the velocity and acceleration emerges during the lowering phase (indicated with grey background in Fig. 1). Indeed, at the beginning of lowering (a), the assistance provided by the *inclination* strategy may be perceived by the wearer as hindering the flexion of the trunk. This behavior is similar to the support of passive exoskeletons, for which the assistance during lifting cannot be incremented at will, as it results in a higher resistance during lowering. However, with the *velocity* and the *acceleration* strategies, the torque working against the subjects can be reduced, correspondingly reducing the resistance for trunk flexion (2). Moreover, it can be observed in Fig. 1 that the box's mass contributes to increase lumbar moment (phases (c) and (d) of lifting and

(a) and (b) of lowering). A solution to account for this need of additional assistance was evaluated in [2]: the assistance is increased when a load is detected on user's hands.

As regards the *acceleration* strategy, one limitation concerns the delay of the acceleration signal (T1-T2 in Fig. 1), as it results in the delay of the dynamic component of the assistance. Furthermore, stability may also be compromised, if the acceleration is overestimated. As this controller acts as compensating the inertia of the user's upper body, overcompensating may lead to feedback inversion and instability [6]. On the contrary, with the *velocity* strategy, force augmentation is achieved by positive feedback (i.e., the generated forces augment the movement initiated by the wearer) that was proved to decrease the user's stability [3]. A further limitation for both strategies, particularly related with real workplace use, is the need to add an IMU on the user's torso to acquire the angular velocity, instead of using the one embedded in the exoskeleton. The need for this sensor is due to the user-exoskeleton coupling, which allows for relative movement between the two, i.e., the motion of the exoskeleton is delayed respect to the user's torso motion.

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