

Enhancing safety and utility of the EduExo Pro exoskeleton for use in a laboratory setting

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Abstract. Robotic exoskeletons are employed in a wide range of applications, from aiding industry workers to post-stroke rehabilitation and support of disabled people in everyday life. Close interaction of exoskeletons with humans imposes several design challenges, one of which is implementing compliant control needed for successful cooperation of the exoskeleton with the user. Additionally, high level of safety is required. This paper presents the implementation of impedance control and gravity compensation in an upper-limb robotic exoskeleton EduExo Pro with the aim of enhancing its safety and utility for use in a laboratory setting. Electronics selection process, mechanical design and software development are described. Additionally, the efficiency of the implemented gravity compensation is evaluated. Results indicate that gravity compensation successfully reduced the average muscle activity, however, further improvement is needed to achieve statistically significant results.

1 Introduction

Hardiman, the first successfully developed exoskeleton prototype, appeared in 1960s. It was designed for military use with the aim of improving the strength and performance of the wearer [1]. Since then, the robotic exoskeletons have proven to be important in a broad area of application spanning from power amplification in an industrial environment to neuromuscular impairment compensation and post-stroke rehabilitation, as well as offering support for disabled people in their activities of daily living [1].

Among the innovators in this field, a Swiss company Auxivio AG has developed a product line called EduExo, which offers a range of educational exoskeletons aimed at making exoskeletons more accessible and enabling students, teachers, hobbyists, and researchers to learn about and teach exoskeleton technology [2]. One of these exoskeletons is the EduExo Pro, which is a robotic exoskeleton with three degrees of freedom (DoFs). The shoulder joint has two passive DoFs, where shoulder flexion is supported by a mechanical spring. The elbow joint has one active DoF driven by a servo motor. Interaction forces with the user are measured with a force sensor, which is installed in the lower arm segment.

Due to the close physical interaction between the exoskeleton and its wearer, a risk of injury can occur in case of unpredicted behaviour of the human or the exoskeleton. Compliance offers to compromise between safety requirements of such interaction and traditional

performance metrics such as tracking accuracy [3]. It is most often implemented in form of impedance or admittance control, which both shape the dynamical relation between actuator velocity (or position) and force [4] and generate compliant motion [5].

Using the hardware provided by the EduExo Pro kit it is possible to implement admittance control of the motor in the elbow joint. Using an admittance model, the desired position can be calculated from the measured interaction force [5]. The performance of admittance control approach, however, depends on the force sensor, which is sensitive to drift and temperature change [6]. Furthermore, if it is not possible to directly control and limit the torque that the motor applies, the exoskeleton could potentially hurt the user during an experiment. Impedance control, on the other hand, implements compliance by using position as the input and the impedance model is used to calculate the output force/torque [5]. In this way, the torque generated by the motor needs to be directly controlled and there is no need for a force sensor. Instead, the motor's encoder can be utilized to measure position, which serves as the input. In contrast to admittance control, impedance control requires compensation of the weight of the robot [7], which can further be extended to arm weight compensation [8].

In this paper we present an upgrade to the EduExo Pro exoskeleton kit for enabling the use of impedance control and the implementation of gravity compensation in the elbow joint.

2 Methodology

2.1 Electronics and mechanical design

To implement impedance control, a motor with current control was needed. We chose the Dynamixel XM540-W270-T/R (Robotis, South Korea) servo motor for its capability of being controlled in position, velocity, or current control mode. The motor is controlled through RS-485 serial communication [9]. To integrate the motor in the exoskeleton we decided to upgrade the processing unit to a Raspberry Pi 4 Model B. The communication between the Raspberry Pi and the motor was enabled by a USB to RS-485 converter (U2D2, Robotis, South Korea).

To be able to implement gravity compensation, a triple axis accelerometer was added to the system (LIS331HH, SEN0407, DFRobot, China).

The system can be powered by a 12 V power supply or 4 cell li-ion battery. Additionally, a DC/DC converter was used to adjust the voltage supplied to the Raspberry Pi to 5 V.

Connections between the electronic components are shown in Figure 1.

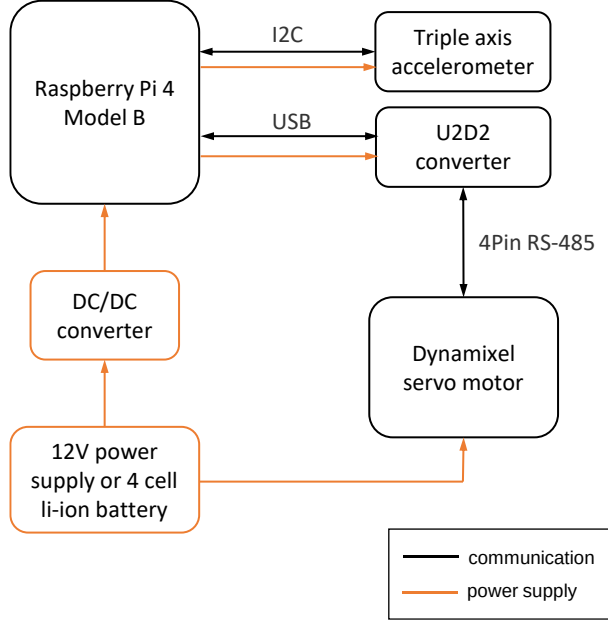


Figure 1: Schematics of electrical components connections.

All the electronic components were placed in the case on the upper arm, which had to be redesigned. Using CAD software, electronic components along with the sheet metal board, which was part of the EduExo Pro hardware, were modelled and a new case was designed. Additional adapters were designed to incorporate the Dynamixel motor to the existing EduExo Pro hardware.

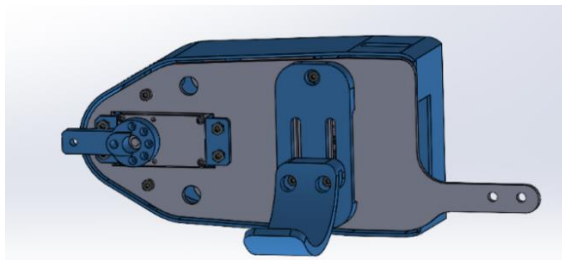


Figure 2: 3D model of the upper arm assembly with newly designed components coloured blue.

To minimise the size of the new case we decided to attach the cuff to the sheet metal board from the outer side, which required the use of an adapter between the board and the cuff. To maintain the distance between the board and the user's arm despite the use of the adapter the cuff had to be redesigned as well. The 3D model of the assembly on the upper arm is presented in Figure 2.

All the of the designed mechanical parts were then 3D printed. Parts that are in contact with the motor were printed from ABS, which can withstand increase in

motor's temperature. Since the motor handle adapter, cuff and cuff adapter are expected to be subject to higher loads, they were printed from carbon fiber reinforced Nylon filament (PolyMide™ PA6-CF, Polymaker). The final design of the exoskeleton is presented in Figure 3.



Figure 3: Subject wearing the final design of the exoskeleton.

2.2 Software

In order to implement impedance control, we set the Dynamixel servo motor in current control mode, where the control input to the motor is the desired current. The mapping between the current and the output torque was modelled based on the supplied performance graph of the motor from the manufacturer. For safety reasons, the maximal generated torque was limited to 8 Nm in our control software as well as by setting a current limit in the motor's internal controller.

The impedance model used is described by

$$\tau = K_s (\theta_{\text{ref}} - \theta_m) + K_d \dot{\theta}_m + \tau_g, \quad (1)$$

where the torque control signal τ is the sum of the 0th order term with the gain K_s , which represents the desired mechanical stiffness, and the 1st order term, where the gain is the damping coefficient K_d . Additionally, we added a term τ_g used for gravity compensation. Moreover, θ_m is the position measured by motor's encoder, θ_{ref} is the reference position, and $\dot{\theta}_m$ is the measured velocity.

The need of implementing gravity compensation is to compensate for the torque caused by gravity force of the mechanical parts of the lower arm segment. Additionally, it offers us the possibility of easily calculating the support required if we want to compensate for the weight of the lower arm of the user.

For this purpose, accelerometer measurements of the rotated gravitational field vector were used. Using the AHRS python toolbox the roll rotation of the upper arm segment around an initially defined earth coordinate system was calculated, describing the shoulder flexion/extension. The rotation matrix $\mathbf{R}_{s,f/e}$ is used to describe this rotation. The flexion/extension in the elbow joint was measured by the motor's encoder and is described by the rotation matrix $\mathbf{R}_{e,f/e}$. In this calculation we assume a vertical orientation of the subject's trunk. Additionally, the shoulder abduction/adduction is

constrained by the exoskeleton and therefore does not influence the level of gravity compensation required.

The rotation matrix $\mathbf{R}_{la}^0$, which describes the rotation of the lower arm coordinate frame relative to the earth coordinate frame, is computed as

$$\mathbf{R}_{la}^0 = \mathbf{R}_{s,f/e} \mathbf{R}_{e,f/e}. \quad (2)$$

This matrix is then transposed to obtain the rotation matrix $\mathbf{R}_0^{la}$, which is used to rotate the gravity vector $\mathbf{f}$ from earth coordinate frame to lower arm coordinate frame as follows

$$\mathbf{f}^{la} = \mathbf{R}_0^{la} \mathbf{f}^0. \quad (3)$$

The torque caused by the gravity vector $\boldsymbol{\tau}_g$ is computed as

$$\boldsymbol{\tau}_g = \mathbf{r}^{la} \times \mathbf{f}^{la}, \quad (4)$$

where $\mathbf{r}^{la}$ is the vector from the motor axis to the centre of gravity of the lower arm presented in the lower arm coordinate frame. The centre of gravity of the user's lower arm was calculated based on anthropometric tables in [10] depending on the user's weight and lower arm length.

The torque which needs to be compensated is the component of the calculated torque around the motor axis.

2.3 Experiment

First, the gravity compensation algorithm was evaluated by monitoring shoulder and elbow rotation angles as well as the compensation torque applied without a human using the exoskeleton. At the start of the experiment the shoulder flexion/extension was 0° (corresponding to upper arm segment being oriented horizontally) and the

elbow joint was at around 60° (0° corresponding to fully extended elbow joint). The shoulder joint was then flexed and extended for around 30° . First, the exoskeleton was controlled using the gravity compensation algorithm described in the previous chapter. Secondly, for comparison, we observed what happened when the same gravity compensation algorithm was used, only considering the elbow joint rotation while the shoulder joint rotations were set to zero.

The efficiency of the gravity compensation was evaluated by surface electromyography (sEMG) measurements of the biceps brachii muscle during the use of the exoskeleton. For this purpose, a Trigno Avanti sensor (Delsys, Natick, MA, United States) was attached to the subject's bicep muscle. 4 subjects were recruited and asked to perform 20 repetitions of elbow flexion/extension movement under each of the three conditions: no gravity compensation, compensation of the weight of the exoskeleton and compensation of the weight of the arm and the exoskeleton. During the experiment with subjects the kinematics was measured by VICON (Vicon Motion Systems, United Kingdom). The kinematics measurements were synchronised with the EMG measurements with a synchronization trigger cable.

The sampling frequency of EMG measurements was 1259 Hz. The acquired data was first filtered using a bandpass Butterworth filter between 20 Hz and 450 Hz. The data was then rectified, the mean of data was subtracted and a lowpass Butterworth filter (10 Hz) was applied to get the signal envelope.

3 Results

3.1 Evaluation of the gravity compensation

Figure 4 shows the results of the first experiment. In the first case, torque oscillations coincide with oscillations of

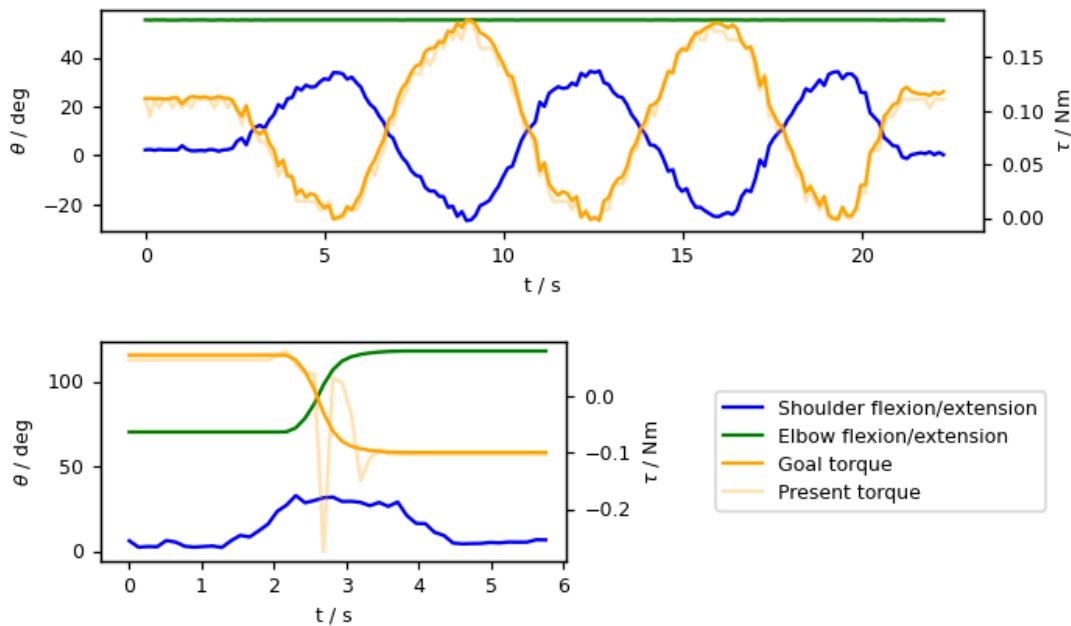


Figure 4: Measured rotations and compensation torque using the implemented gravity compensation model (above) and using the gravity compensation model which only considers the elbow rotation (below).

the shoulder flexion/extension, as expected. The minimums of the applied torque value occur when the shoulder flexion/extension takes the maximum value. It can be observed that at this moment, shoulder and elbow flexion/extension angles add up to around 90°, which coincides with lower arm segment being positioned vertically. Elbow flexion/extension, on the other hand, is constant throughout the experiment. This shows that the gravity compensation algorithm is supplying the appropriate amount of torque.

In the second case, however, shoulder rotation caused a rotation in the elbow joint, which is expected since the compensation torque calculation, in this case, does not consider shoulder rotations. When shoulder joint rotation approached zero again, however, the elbow rotation became again constant.

3.2 Experiment with subjects

To evaluate the efficiency of gravity compensation method the EMG measurements had to be evaluated. For this purpose, time windows were defined, which start once the angle of the elbow joint reaches 25° (relative to the fully extended arm) and are 1 s long, which roughly corresponds to the point in which subjects fully flexed the elbow joint. Average and standard deviation values of the voltage measured during the time windows for each subject are shown in Figure 5. Additionally, average values for all subjects under a certain condition are shown.

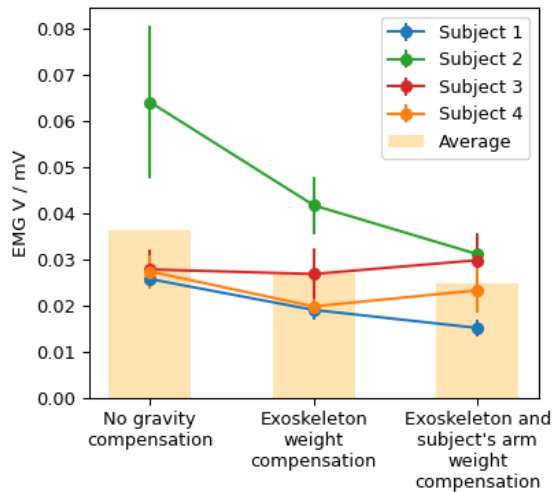


Figure 5: Exoskeleton user's muscle activation with different levels of gravity compensation provided.

Results show that three out of four subjects experienced decline in muscular activity with the introduction of gravity compensation. The level of decrease depends on the subject. When the arm weight compensation was added the muscular activity further decreased in two subjects but increased in the other two subjects. A repeated measures ANOVA test was conducted to examine the effect of gravity compensation on mean EMG value during elbow flexion for the four subjects. To achieve normal distribution, logarithmic transformation of the data was used. The test showed no

significant difference between the EMG values measured without gravity compensation, with exoskeleton weight compensation and with exoskeleton and arm weight compensation with $F(2,6) = 3.66$, $p = 0.09$.

4 Conclusion

In this paper the process of implementing impedance control and gravity compensation for the EduExo Pro exoskeleton is presented. The preliminary results showed that the exoskeleton can correctly compensate its own weight. However, the efficiency of the exoskeleton for arm weight compensation was not confirmed with the EMG measurements. One possible cause for this might be an improper fit of the device on the subjects, causing misalignments between the joint of the exoskeleton and the joints of the subject. This could have introduced additional forces on the subject which resulted in increased muscle activity. Another possibility is that there was a small effect due to friction in the gearbox of the motor. In future work, the utility of the exoskeleton could therefore be further improved by implementing software compensation of residual friction torques of the actuator gearbox.

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