Development of a Modular Test Bench for Cable-Driven Synergy-Based Exosuit Actuation and Control Strategies

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Abstract. In their survey dated 2019, the UN predicts that the population above the age of 65 will increase from 9%, to 16% by the year 2050 [1]. Ageing results in the eventual loss of muscle mass and strength, joint problems and overall slowing of movements with a greater risk of suffering falls or other such accidents. Gait assist exoskeletons can help promote active ageing in this segment of the population. Since these devices are user specific in terms of the mechanics and control required, the facility to test different parameters becomes indispensable. This work details the design and construction of a modular test bench to implement synergies in an exosuit using motors and cables, and the optimization of the control scheme to better adapt it to specific patients.

Keywords: Exosuit · Modular Test Bench · Synergies

1 Introduction

Wearable exoskeletons, also known as 'exosuits' are relatively recent. One of the first works dealt with lower limb exosuits [2, 3] with textiles for force transmission. These exoskeletons actuate the hip and ankle, using force sensors on the sole of the foot to detect gait phases. Position control based on a predetermined gait was used to drive the motors. These designs allowed a metabolic cost reduction of 6.4% to 19% over several iterations [4]. The XoSoft EU project [5] created an exosuit for people with mobility problems. A prototype beta 1 used an elastic band and clutch in series. Another exosuit design [6] used strain gauges and IMUs to measure the angles and a kinematic model to estimate the gait cycle from these measurements. It used FSRs on the soles of the feet to detect the phase of the gait cycle. Alternatively, there are exosuit designs [7] controlled by PD control with iterative learning (ILC) and able to achieve 15.67% metabolic cost reduction. A recent design [8] features an exosuit that weighs only 1.8 kg. The gait cycle was detected with IMUs as well as measured forces and position by the motor encoders. The trade-off of this weight reduction is a reduction in force transmission (between 50–100 N) compared to other designs, but it still achieved an 11.52% reduction in metabolic cost.

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2 Methodology

Weight reduction in such devices is primordial. One such concept used to characterize the largest possible number of movements using the least number of actuators is that of synergies [9, 10]. To characterize these synergies, both the mathematical models and the physical version play important roles in verifying that the system functions as expected. In order to find the optimal actuation scheme, it is necessary to perform several tests, modifying design parameters to see which of them is the most suitable for the needs of this project, to appropriately assist the user's gait. Therefore, this work presents the design and control of a modular test bench that will allow testing different types of actuation systems, control strategies and assistance configurations (ankle only, knee only, hip only or any combination of them) allowing easy assembly and disassembly for rapid prototype development.

2.1 Test Bench Design Theory

The basis for the design of the exosuit based on synergies are the torque curves (Fig. 1) of each leg segment. The torque curves were calculated based on a mathematical model with information from a public database [11], and anthropometric data [12]. The goal is to assist the three segments of the leg with the least number of actuators. Based on the assumption that the exosuit only assists at specific times in the gait cycle, the objective is to assist the subject with the torque necessary to assist a certain percentage of the total joint torque.

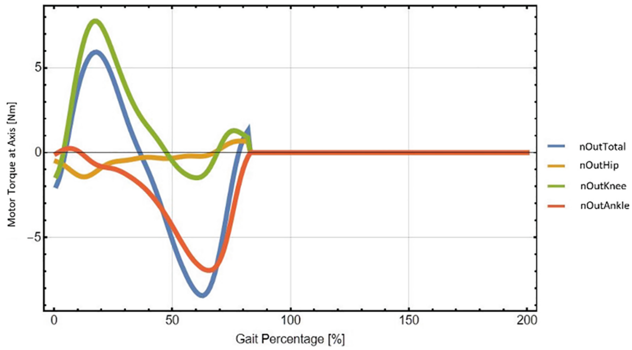

Fig. 1. Joint Torques. Blue: Total. Yellow: Hip. Green: Knee. Red: Ankle.

Observing the graph in Fig. 1 of joint torques, it can be seen that the maximum torque at the hip and ankle is positive at various points of the gait (extension). Due to cable actuation being restricted to extension only, areas where gait assistance is viable are in the negative region (flexion). Moreover, the torque required at the lower leg or knee joint is negative only during a small portion at the end of the gait. Therefore, torque can be supplied to the knee at the end of the cycle to assist gait and achieve a further reduction in metabolic cost. The mathematical model used is detailed in [10]. Thus, it was decided to actuate the hip and ankle at the same time and the knee at the end of the cycle. Two motors with different pulleys are used for this purpose. One motor has a two-pulley train and the other is a three-pulley train (although only two are used in this trial taking into account the decision taken previously) designed based on subject parameters from the database. The first motor acts according to its own gait curve, generated with the model, controlling the amount of cable extension, winding or unwinding the cable as appropriate to generate the synergy from the tension at the anchor points. The second motor may be activated to restrict the extension, and held still on another run to compare the resulting curves to see the effects of the synergy.

For this purpose, the desired curve of each motor is processed and recorded in the motor controllers. The motors are controlled by a PID controller. To estimate the position values to send to control the motor in angular position, the expression used:

$$P_{desired} = \frac{2^{n_{encoder}} \times G_A \times \omega_{req}}{2\pi \times G_B} \tag{1}$$

where $P_{desired}$ is the position to be sent, $n_{encoder}$ is the number of bits of the motor encoder, G_A and G_B are the reduction coefficients of the motor gearbox, ω_{req} is the desired angular position. A similar equation is used for displacement curves. The resulting cable displacement is calculated from load cell readings, the motor encoders and a linear sensor for accuracy. The tracking error was calculated by comparing the position detected by the encoder versus the desired position. The displacement measured by the motor encoder is calculated knowing the pulses per revolution p_{enc}:

$$d_{cable}^{encoder} = \frac{n_{pulses} \times 2\pi\, r_{pulley}}{(2)^{p_{enc}}} \tag{2}$$

where n_{pulses} is the number of pulses from the encoder, r_{pulley} is the pulley radius.

In this setup, springs of stiffness k = 1060 N/m were used. Each spring had a maximum displacement of 29.7 mm, so 3 springs were used in series to obtain a total displacement of up to 9 cm. With this range, the scaling factor of the input curves could be as little as half the values needed in the exosuit.

2.2 Construction of the Test Bench and its Modules

Each part was designed and then assembled virtually in CAD to verify compatibility before being assembled on the physical bench. Pulleys and brackets were 3D printed in high-strength PLA 870 filament. The test bench itself is made up of perforated steel plates (R4T6 of 1000 mm × 500 mm) for versatility. The motors used in the test bench are Maxon EC 4-pole (BLDC/brushless) 200 W, with a maximum torque of 95 mNm. In order to increase the torque, two different types of gearboxes were mounted: one 33:1, and the other 79:1, with a resulting torque of 3.13Nm and 7.5 Nm respectively. Both motors have 12/20-bit multiturn absolute encoders for feedback to implement position control of the motors. The motor controllers are the Maxon EPOS4 50/8, to which both the Hall effect sensors of the motors and the absolute encoder are connected. This controller is connected to the PC via USB COM or to the Arduino via RS232, following the object dictionary of the EPOS4. Two types of cables were used as tensile elements for the tests: one of the cables was a 1.5 mm steel cable consisting of 6 ropes, 7 wires and a load of up to 26 kg with a safety factor of 5:1. The other was a 1.5 mm bicycle brake cable with its corresponding sheath. For the measurement of forces and stresses, Futek S-beam load cells (LSB201) were used, connected to an appropriate amplifier, capable of measuring a maximum load of 445 N. The linear displacement sensor consists of a potentiometer connected to the analog to digital converter (ADC). The travel of the potentiometer is 100 mm, and it has a proportional voltage output. This allows for cable displacement feedback without having to use a very high scale factor, economically and without overcomplicating the design. For the sensor connection interface, a 32-bit Arduino DUE was used. This system was also used to send commands to the motors and to control the clutches.

3 Tests Conducted

3.1 Motor Position Control

The first test consists in assembling the motor with its support, the pulleys or other mechanical parts except the cables. The goal of this experiment is to optimize the operating parameters of the position control system by accounting for the impedance measured in the transmission system to be used.

3.2 Analysis of Synergy Based Actuation Strategies

The next test consists of applying the gait curves obtained by means of synergies to record and evaluate the assistive capacity of the exosuit.

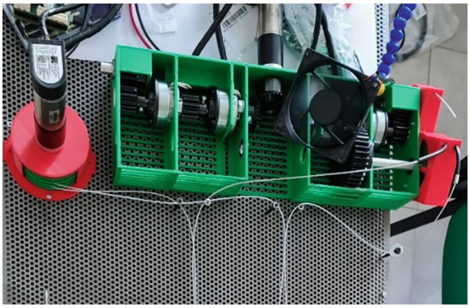

Fig. 2. Dual Motor Synergy Test Setup

This test consists of mounting synergy-based pulley trains designed on the basis of the hip and ankle synergies on each motor and passing the cable between them as in Fig. 2. To record the extension of the cables, load cells are used at the junction points with the corporate segment, using one for each segment to be measured.

4 Results and Discussion

4.1 Adjusting Motor Position Control Parameters

The following graph shows the curve sent to the motor (blue), the position measured through the encoder (red) and the tracking error (green).

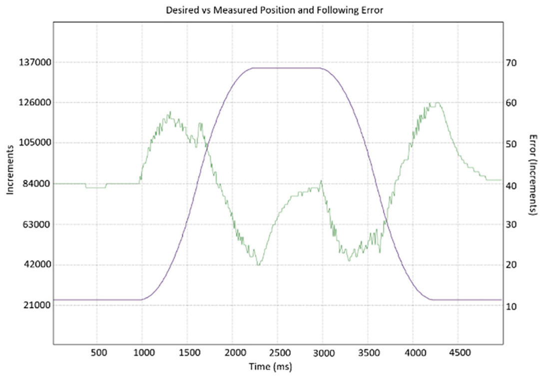

Fig. 3. Tuning the Position Control Algorithm. Blue: Desired Position. Red: Measured Position from Encoder. Green: Error.

As can be seen in Fig. 3, the motor follows the position curve with a high level of accuracy due to the PID controller tuning. After a simple manual optimization, even with a very high acceleration of 4297 RPM/s and a speed of 10000 RPM the maximum position error measured was 60 increments, which corresponds to an angle error at the pulley shaft exit of 0.639°. The results obtained are more than sufficient for present requirements allowing tests to be performed with good accuracy.

4.2 Analysis of Synergy-Based Actuation

As expected, due to the lack of tension on both pulleys at the pulley output due to the current design of the cable feed system, differences in the shape of the tension curves compared to the position curves are observed. Figure 4 shows lack of tension in the ankle cable observable from samples 100 to 170 approximately, where the motor rotates but does not displace the ankle cable correctly. This limitation will be resolved in the future by including a feeder design for the cables to stay properly tensioned.

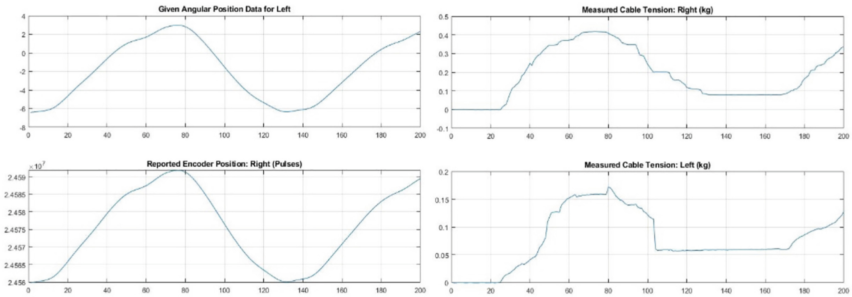

Fig. 4. Synergy Test Results: Two Motors for Two Segments.

5 Conclusions

The bench has been indispensable in the optimization of system parameters for adjusting motor control system parameters, estimating actuation times, deciding the sizes of various system components, etc. to facilitate testing of the actuation systems in exosuits, reducing prototyping and construction costs. A potential improvement is the coupling of a third motor in place of the springs in order to better simulate non-linear dynamics at the output. This motor can be fed with curves that represent some gait pathology such as weak muscles or pathological gait, in order to design, improve and measure how the control system adapts to these irregularities.

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