Design, control and evaluation of a low-cost active orthosis for the gait of spinal cord injured subjects

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Robotic gait training after spinal cord injury is of high priority to maximize independence and improve the living conditions of the patients. Current rehabilitation robots are expensive and heavy, and are generally found only in the clinical environment. To overcome these issues, we present the design of a low-cost, low-weight and personalized robotic orthosis for incomplete spinal cord injured subjects. The paper also presents a preliminary experimental evaluation of the assistive device on one subject with spinal cord injury that can control hip flexion to a certain extent, but lacks control of knee and ankle muscles. Results show that gait velocity, stride length and cadence of walking increased (24,11%, 7,41% and 15,56%, respectively) when wearing active orthoses compared to the case when the subject used the usual passive orthoses.

1. Introduction

Spinal cord injury (SCI) is prevalent in society. Worldwide each year more than 250 000 individuals become spinal cord injured; with road traffic crashes, falls and violence as the three leading causes [1]. Walking impairment after SCI leads to a decreased quality of life and other serious health conditions, e.g., chronic pain, vein thrombosis, urinary tract infections, pressure ulcers, or respiratory complications among others; and it also carries substantial individual and societal health care costs. Among the essential measures for improving the survival, health, and social inclusion of people with SCI, the World Health Organization (WHO) recommends improving access to skilled rehabilitation services to maximize function, independence, overall well-being and community integration; and to appropriate assistive devices that enable people to perform everyday activities, reducing functional limitations and dependency [1]. Locomotor rehabilitation has been reported as a high priority issue for subjects with SCI independent of severity, time after injury, and age [2]. And, furthermore, robotic actuation has recently shown to be useful (and effective when combined with manual therapy) for neurorehabilitation and lower limb motor function recovery [3].

Current gait rehabilitation robots are machines that have been conceived without taking into account the patient's specific anatomy and actual degree of disability. These gait trainers can be machines that partially or totally support the patient's weight and train the walking motion over a treadmill or feet supports [4-6]; or lower limb exoskeletons that assist over-ground walking so that the patient bears his or her weight [7-9]. The first group of robots are heavy and expensive, and they are generally only found in hospitals and rehabilitation centres. However, the second group are less heavy and costly, but so far they have been only used in health care centres because skilled personnel have to manually fit the robot to the patient and operate the exoskeleton. The main drawbacks of the current technologies are: (i) due to their cost, weight and operating complexity, these machines are only found in health care centres and technical qualified personnel are required for operation and supervision; (ii) they are adapted to the patient directly at the health care centre, increasing treatment time and therefore health care costs; (iii) in general, their functional approach is to impose a movement rather than to cooperate with the patient; and (iv) in developing countries, even hospitals or rehabilitation centres have difficulty acquiring such high-cost technology.

In the last years a number of robotic prototypes aimed at assisting or rehabilitating gait have been developed in research labs [10]. For example, active ankle-foot orthoses (AFO) to treat individuals with drop-foot gait [11]; active knee-ankle-foot orthoses (KAFO) for subjects with more severe gait dysfunctions [12]; and the stance-control KAFO (SCKAFO), a device that incorporates a locking system to constrain knee flexion during the stance phase [13]. In the latter, some systems are based on mechanisms that lock flexion at a fixed position, others on electromagnetic wrap-spring clutches and, finally, other systems are based on friction. To assist the gait of complete paraplegic patients, the hip joint must also be actuated. This is the case of the active exoskeletal devices presented in [14-16].

This work presents the design, control and experimental evaluation of a novel robotic orthosis for gait assistance in subjects with SCI. More precisely, the prototype is intended for patients that can control hip flexion to a certain extent, but lack control of knee and ankle muscles. With the aim of obtaining a low-cost, low-weight and simple assistive device, the design is based on the standard passive knee-ankle-foot orthoses that these patients use after rehabilitation at the health care centre. The latter include a knee locking system, which is essential to bear the patient's weight during stance due to the lack of quadriceps force; and a compliant system that applies a dorsiflexion torque at the ankle to avoid drop-foot gait (*klenzak* joint). The active orthosis presented here improves the passive devices by adding a motor at the knee, that can move or lock the joint, and an inertial measurement unit (IMU) at the shank to detect gait events. The aim of this work is two-fold: first, we present the mechanical design and the control architecture of the robotic orthosis; and second, we perform a preliminary experimental evaluation on a subject with SCI. In this pilot study, the kinematics of walking with passive orthoses is compared with that obtained with the presented robotic device.

2. Mechanical design

The proposed lower limb active orthosis has two degrees of freedom (DOF). The knee joint is powered through an electrical motor in series with a Harmonic Drive gearbox, and the ankle is passively actuated through a compliant mechanism that applies the above-mentioned dorsiflexion torque (*klenzak* joint). This exoskeletal device is conceived for over-ground gait assistance at home or in a clinical environment. Preliminary designs where reported in [17], which presents a CAD design of a prototype that uses a motor with a planetary gearbox, a controllable locking mechanism at the knee, and an ankle encoder and on-off contact sensors for control; and [18], which presents different versions of the same product, being the last version a prototype including a motor with a planetary gearbox at the knee, and an ankle encoder and force-sensing resistors at the foot-ground interface for control.

The current device weights 2.7 kg per leg, along with a 1.7 kg backpack containing a BeagleBone Black board, the motor drivers and the battery. The bilateral thigh and shank uprights are articulated at the knee, using a standard hinge joint at the medial side and the motor-gearbox set at the lateral side. A footplate with a shoe is hinged to the shank uprights by the compliant *klenzak* joint, which allows an ankle range of motion between 10º (plantarflexion) and 15º (dorsiflexion). The initial pre-tension of the joint spring can be adjusted to the patient needs. It is important to bear in mind that the orthosis structure is specifically tailored to the patient to avoid the problem of adapting the same design to the wide range of morphologies found among patients with SCI. This is in fact the current process made to adapt passive orthoses at the orthopaedics workshops. Figure 1(a) shows the right robotic orthosis with the elements described later.

(a)   (b)

Motor with
Harmonic Drive

Inertial measurement
unit (IMU)
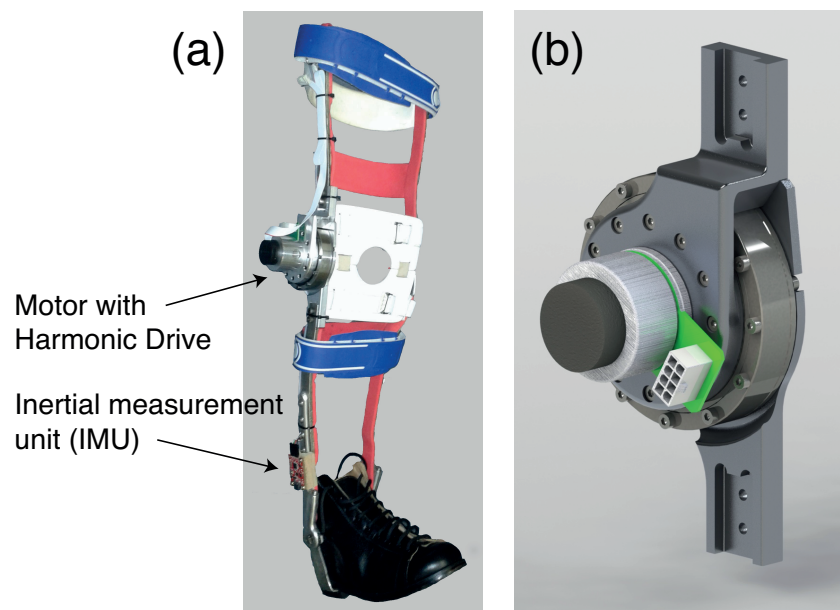
Figure 1: Robotic orthosis design: (a) general view showing the knee actuation system and the inertial measurement unit (IMU); (b) CAD design of the developed actuation system.

The design and selection of the orthosis actuation system were based on kinematic and kinetic data of the knee joint (angular velocity, torque and power) during normal gait at a normal speed [19]. The most significant criteria for the actuation system selection were the specific power (power to weight ratio of the actuation system), the system dimensions and the portability of the power supply system.

Based on the considerations above, a 70 W brushless DC motor (EC45 flat, Maxon Motor AG, Sachseln, Switzerland) was selected, which has a nominal voltage of 24 V and a nominal torque of 128 mNm (maximum continuous torque). A Harmonic Drive gearbox (SHD-20-160-2SH, Harmonic Drive AG, Limburg-Lahn, Germany) is coupled to the motor to increase torque and reduce velocity, which offers a large gear ratio with a reduced space. In this particular case, the selected gear ratio of 160:1 allows a continuous net torque at the knee of 20,5 Nm and peak torques of 60 Nm according to the motor driver current limit. Figure 1(b) shows the CAD design of the developed actuation system.

3. Sensors and control algorithm

All the sensors are placed on the orthosis mechanical structure in order to avoid issues related to safety, comfort, reliability and donning/doffing process [16]. The sensors used are one IMU and one angular encoder coupled to the knee motor per orthosis, see figure 1(a). The low-cost 9-DOF IMU (SparkFun Electronics, Niwot CO, USA) is attached to the shank upright; and incorporates a triple-axis gyro, a triple-axis accelerometer and a triple-axis magnetometer. The orientation and acceleration measurements are sent to the BeagleBone board through a serial interface.

The control algorithm is implemented in two layers. The internal layer consists of a PID controller that keeps the leg straight during the stance phase, and performs a predefined flexion-extension cycle during the swing phase. In order to detect when the swing cycle must be triggered, an outer algorithm based on the IMUs measurements is used.

3.1. Knee angle control

The knee angle θ is kept equal to zero (i.e., straight leg) during the stance phase of walking, and is set to track a predefined trajectory during the swing phase, such that:

$$\theta(t) = \frac{k_a}{2}\left[1 - \cos\left(\frac{2\pi}{t_c}t - \phi(t)\right)\right], \ 0 \leq t \leq t_c \tag{1}$$

where k_a is the maximum knee flexion angle, t_c is the cycle duration, and $\phi(t)$ is a phase angle that tunes the cycle by deforming the shape of the $\theta(t)$ curve:

$$\phi(t) = k_s \sin\left(\frac{\pi}{t_c}t\right) + k_w \sin\left(\frac{2\pi}{t_c}t\right) \tag{2}$$

In equation (2), parameter k_s slants the peak forward or backwards, thus modifying the relative duration of flexion and extension, whereas k_w increases or decreases the peak width. The four parameters defining the curve (t_c, k_a, k_s, k_w) can be modified from the user interface in real time, in order to better personalize the cycle to the gait pattern of the patient.

The knee motor follows the predefined trajectory by using the EPOS2 (Maxon Motor AG, Sachseln, Switzerland) implementation of the so-called interpolated position mode (IPM), defined according to the CANopen standard CiA® 402 V3.0. The controller receives a list of PVT (position, velocity, time) vectors, and performs a cubic spline interpolation between them. The EPOS2 internal controller tracks the interpolated trajectory by means of a PID algorithm with feedforward, cascaded with a PI controller that sets the motor current.

3.2. Swing detection state machine

Each shank incorporates an IMU that provides its orientation, linear acceleration and angular velocity at a 100 Hz rate. The algorithm detects the gait intention by relying exclusively on these inertial sensors, triggering the flexion-extension cycle when the following four conditions are met:

- The vertical acceleration in the ascending direction overcomes a trigger value.

- The vertical acceleration has remained within a threshold for at least a time interval.

- The shank has a minimum forward inclination angle.

- The opposite shank has a minimum backwards inclination angle.

The second condition ensures that the foot has been resting on the ground just before the trigger occurs. The third and fourth conditions are safety checks: on the one hand, they prevent the cycle from being launched when the patient raises a foot in standing position; and, on the other hand, the angles can be configured in a restrictive way to prevent false accelerometer triggers when a foot is on the ground.

4. Experimental evaluation

The device was tested on an adult female 41 years old, mass 65 kg and height 1.52 m; with incomplete spinal cord injury at T11. Before the tests, the patient was able to walk with her passive knee-ankle-foot orthoses, that included the knee locking system and the compliant ankle joint.

Two experiments were performed in this preliminary evaluation. In experiment 1, the patient walked with her usual pair of passive knee-ankle-foot orthoses with the help of two parallel bars. Then, the subject carried out six one-hour training sessions wearing the active orthoses and did some specific exercises at home to facilitate adaptation to the new assistive device. After this period, a second experiment (experiment 2) walking with the developed active orthoses was performed as seen in figure 2(a). In this case, the patient also walked with the help of the two parallel bars.

In order to compare the walking kinematics in the two experiments, four consecutive gait cycles were captured each time by six optical infrared cameras (Flex:V100, Natural Point, Corvallis, USA) that measured the three-dimensional position of 37 reflective markers. Then, a computational 3D skeletal model with 18 anatomical segments and 57 degrees of freedom was used to determine different kinematic characteristics of the subject's gait, see figure 2(b). In this study, we compared gait velocity, stride length, cadence of walking, and centre of mass (COM) lateral displacement between experiments 1 and 2.
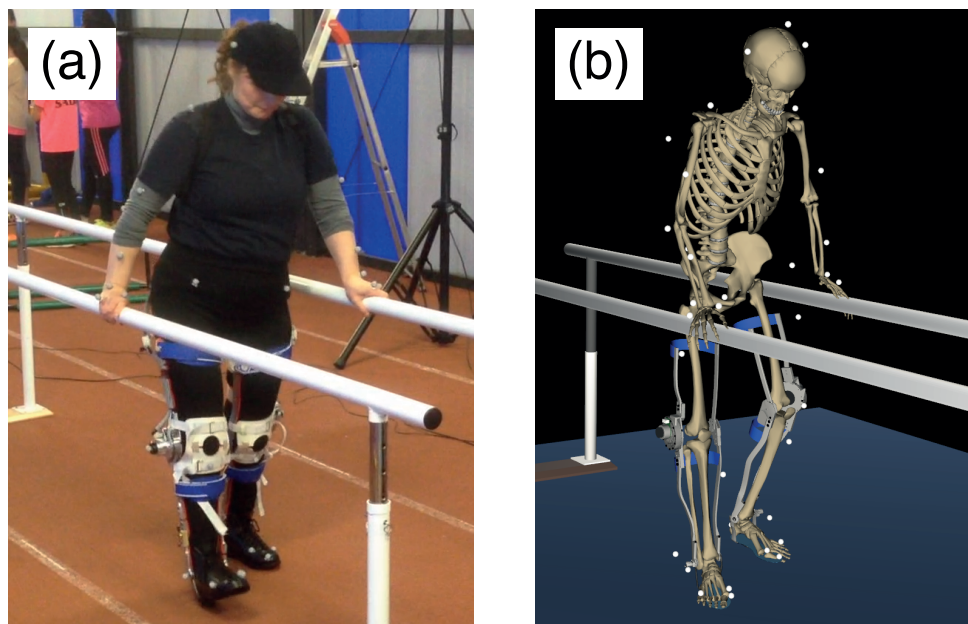
Figure 2: Gait of spinal cord injured subject assisted by active orthoses and parallel bars: (a) acquired motion; (b) computational model.

Table I shows the above-mentioned kinematic descriptors for one gait cycle during experiment 1 and another gait cycle during experiment 2. It can be seen that gait velocity, stride length and cadence of walking increased (24,11%, 7,41% and 15,56%, respectively) when wearing active orthoses compared to the case with passive orthoses. Furthermore, the lateral displacement of the subject's COM decreased in 19,31% when the subject walked with active orthoses. These results indicate that the gait obtained when using the active orthoses was faster and more balanced.

Table 1: Kinematic data with passive and active orthoses

	Experiment 1: Passive	Experiment 2: Active	% change
Gait velocity (m/s)	0,17	0,21	+24,11
Stride length (m)	0,53	0,57	+7,41
Cadence (step/min)	38,46	44,44	+15,56
COM lat. displ. (cm)	7,89	6,37	−19,31

5. Conclusions

This paper presents the mechanical design and control of a patient-tailored, low-cost and low-weight knee-ankle-foot robotic orthosis for subjects with SCI. This orthosis is equipped with a compact knee actuation system, composed by an electrical motor and a Harmonic Drive gearbox, and an inertial measurement unit at the shank to detect gait events. The system is portable by means of a backpack worn by the user containing a BeagleBone Black board, the motor drivers and the battery.

This work reports preliminary experimental evaluation of the presented assistive device on a female subject with incomplete SCI. Motion analysis at the kinematic level shows that the subject walked faster, and in a more balanced and stable way when wearing the active orthoses as compared to the case with passive orthoses. While the experiments provided promising results, more tests with a larger sample of subjects are needed in order to confirm the improvements when walking with the designed orthosis.

Future work will be devoted to improving the existing device, by increasing its robustness, portability, usability and efficiency (reduction of energetic consumption). The latter will be approached by including elasticity in series with the knee actuator to store and release energy during the walking cycle, and improve the average electrical efficiency of the motor during its working cycle. Finally, the inclusion of functional electrical stimulation (FES) to the device and the study of motor-FES co-actuation are also considered as future lines of research.

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7. Referencias

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