ABLE: Assistive Biorobotic Low-cost Exoskeleton

Josep M. Font-Llagunes[1], Alfons Carnicero-Carmona[1], Joan Vidal[2], Daniel Clos[1], Urbano Lugrís[3], F. Javier Alonso[4], Javier Cuadrado[3]

[1] Universitat Politècnica de Catalunya, Spain, josep.m.font@upc.edu
[2] Institut Guttmann, Spain, jvidal@guttmann.com
[3] University of La Coruña, Spain, javicuad@cdf.udc.es
[4] University of Extremadura, Spain, fjas@unex.es

Abstract

Robotic gait training after spinal cord injury is of high priority to maximize independence and improve the health condition of these patients. Current rehabilitation robots are expensive and heavy, and are generally found only in the clinic. To overcome these issues, we present the design of a low-cost, low-weight, personalized and easy-to-use robotic exoskeleton for incomplete spinal cord injured subjects based on simple modular components that are assembled on the current passive orthopedic supports. The paper also presents a preliminary experimental assessment of the assistive device on one subject with spinal cord injury that can control hip flexion, 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 the robotic exoskeleton compared to the case when the subject used the usual passive supports.

Keywords: Robotic Exoskeleton, Human Gait, Spinal Cord Injury, Rehabilitation Robotics, Biomechanics.

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 [2]. 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, respiratory complications or pressure ulcers among others. Moreover, it also carries substantial individual and societal healthcare 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 social inclusion; and to appropriate assistive devices that enable people to perform everyday activities, reducing functional limitation and dependency [2]. Locomotor rehabilitation has been reported as a high priority issue for subjects with SCI independent of severity, time after injury and age [7]. Furthermore, robotic actuation has recently shown to be useful (and effective when combined with manual therapy) for neurorehabilitation and lower limb motor function recovery [14].

Current gait rehabilitation robots are machines that partially or totally support the patients' weight and train the walking motion over a treadmill or feet supports [6,13,15], or lower limb exoskeletons that assist over-ground walking so that patients bear their own weight [9,12,16]. The first group of robots are heavy and expensive, and are generally only found in the clinic. However, the second group are less heavy and costly, but so far they have been only used in healthcare centres because skilled personnel have to manually fit the robot to the patient and operate the exoskeleton. Even so, their price is still out of the reach for the majority of the SCI population. The main drawbacks of the current technologies are: (i) due to their cost, weight and operating complexity, these machines are only found in healthcare centres and technical qualified personnel are required for operation and supervision; (ii) they are adapted to the patient directly at the healthcare centre, increasing treatment time and therefore healthcare 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 difficulties for 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 [8]. For example, active ankle-foot orthoses (AFO) to treat individuals with drop-foot gait [3]; active knee-ankle-foot orthoses (KAFO) for subjects with more severe gait dysfunctions [1]; or the stance-control KAFO (SCKAFO), a device that incorporates a locking system to constrain knee flexion during the stance phase [19]. In the latter, some systems are based on mechanisms that lock flexion at a fixed knee angle, 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 [4,17,18].

This work presents the design, control and pilot experimental assessment of a novel robotic exoskeleton 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 supports that these patients use after rehabilitation at the healthcare centre. These supports 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 limits ankle plantarflexion, thus avoiding drop-foot gait. A modular actuation and sensory system is added to the passive supports providing the needed assistance, based on a motor at the knee, which can move or lock the joint, and an inertial measurement unit (IMU) at the shank to detect gait events and user intention.

The aim of this work is two-fold: first, we present the mechanical design and the control architecture of the robotic exoskeleton; and second, we report some kinematic outcome measures after a preliminary experimental assessment of the exoskeleton on a subject with SCI. More precisely, in this pilot study the kinematics of walking with the robotic device is compared to that when the patient uses the usual passive supports.

2 EXOSKELETON DESIGN

The ABLE exoskeleton brings a significant change compared to the current assistive technologies due to its simple and modular approach, resulting in a low-cost, low-weight, personalized and easy-to-use device that allows to continue rehabilitation outside of the clinical setting. This, in turn, promotes patient's independence and empowerment. The developed device consists of three modular components: an actuation system at the knee that acts as an external muscle, a sensor at the shank that detects the user intention, and a backpack containing the electronics and power supply. The actuation and sensor modules are installed in the current passive orthopaedic supports, which are often owned by the patients with SCI that cannot move the knee and the ankle.

2.1 KNEE ACTUATION MODULE

The proposed lower limb robotic exoskeleton 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 joint that limits plantarflexion. This exoskeletal device is conceived for over-ground gait assistance at home or in a clinical environment. Preliminary designs were reported in [10], 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. Different versions of the same product were presented in [11], 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 electronic 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 is hinged to the shank uprights by the mentioned compliant joint. Finally, a pair of sport shoes are placed on the subject's feet, outside the footplates. It is important to bear in mind that the exoskeleton 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 supports at the orthopaedics workshops. Figure 1 shows the right robotic exoskeleton with the elements described later.

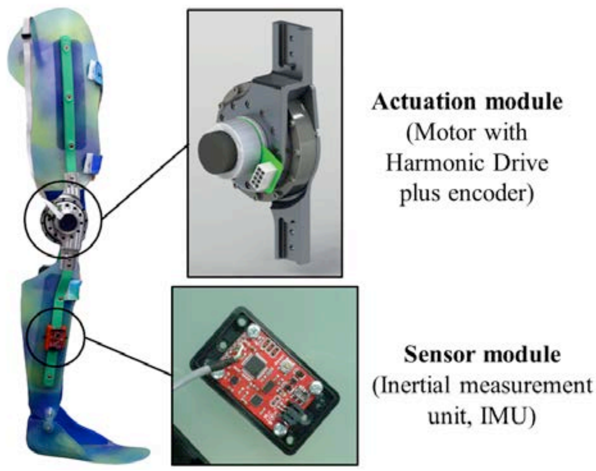

Figure 1: Robotic exoskeleton design: general view showing the actuation and sensor modules.

The design and selection of the exoskeleton 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 [5]. The actuation system was selected taking into account the specific power (power to weight ratio of the actuation system), the system dimensions and the portability of the power supply.

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. Finally, one angular encoder is coupled to the knee motor to measure its angular position. Figure 1 shows the CAD design of the developed actuation module.

2.2 SENSOR MODULE

All the sensors are placed on the exoskeleton mechanical structure in order to avoid issues related to safety, comfort, reliability and donning/doffing process [4]. The sensor module consists of a low-cost 9 DOF IMU (SparkFun Electronics, Niwot CO, USA) enclosed in a protective plastic case that is attached to the shank upright. The IMU 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 using Wi-Fi communication.

The control algorithm is implemented in two layers. First, an internal layer consisting of a PID controller keeps the leg straight during the stance phase, and performs a predefined flexion-extension cycle during the swing phase. Second, an outer algorithm based on the IMUs measurements detects the time when the swing cycle must be triggered based on user stepping intention.

2.2.1 Knee angle control

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

$$\mu_A(x) = \exp\left(\frac{-(x-\mu_A(x))^2}{2\sigma_A{}^2}\right) \quad (1)$$

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 θ(t) curve:

$$\mu_A(x) = \exp\left(\frac{-(x-\mu_A(x))^2}{2\sigma_A{}^2}\right) \quad (2)$$

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 flexion-extension cycle to the gait pattern of each 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.

2.2.2 Swing detection state machine

An IMU, which is attached to the shank of the exoskeleton structure, provides its orientation, linear acceleration and angular velocity at a 100 Hz rate. The algorithm to detect the gait intention relies 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, meaning that stance phase has occurred. 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.

2.3 POWER SUPPLY

The exoskeleton is powered by a compact lithium polymer (LiPo) battery pack with six cells in series giving a nominal voltage of 22.2 V (direct current) and a capacity of 4500 mAh. As mentioned, the battery pack is placed inside the backpack worn by the subject and it powers two motors, one per leg, the motor drivers plus the BeagleBone board, which is powered with 5 V using an adjustable switching regulator (PTN78020W, Texas Instruments, Dallas TX, USA), see Figure 2. The IMUs are directly powered by the BeagleBone board.

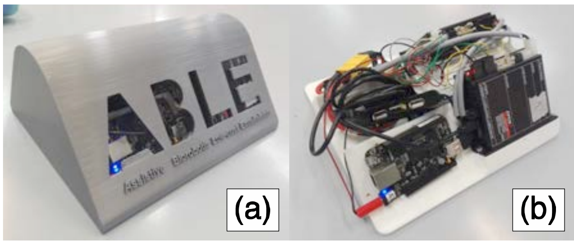

Figure 2: Inside of the backpack containing the electronics: (a) with protective plastic case; (b) without protective plastic case: battery pack, motor drivers, BeagleBone board and switching regulator.

3 EXPERIMENTAL ASSESSMENT

The experimental assessment was performed using a previous prototype of the exoskeleton developed by the group. 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, who can control hip flexion to some extent, but has not control over knee and ankle muscles. Before the tests, the patient was able to walk with her passive knee-ankle-foot supports, which included the knee locking system and the compliant ankle joint.

Two experiments were performed in this preliminary assessment. In experiment 1, the patient walked with her usual pair of passive supports with the help of two parallel bars. Then, the subject carried out six one-hour training sessions wearing the robotic exoskeleton and did specific exercises at home to facilitate the adaptation to the new assistive device. After this period, a second experiment (experiment 2) in which the subject walked with the developed robotic exoskeleton was performed, as seen in Figure 3(a). In this case, the patient walked with the help of two parallel bars as well.

In order to compare the walking kinematics during 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 outcome measures of the subject's gait, see Figure 3(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.

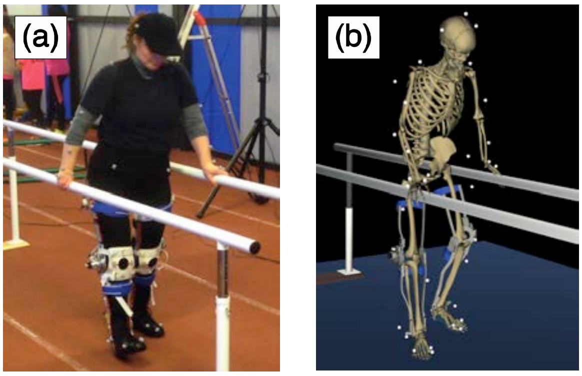

Figure 3: Gait of spinal cord injured subject assisted by robotic exoskeleton and parallel bars: (a) acquired motion; (b) computational model.

Table 1 shows the above-mentioned kinematic descriptors for one gait cycle during experiments 1 and 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 the presented robotic exoskeleton compared to the case with passive supports. Furthermore, the lateral displacement of the subject's COM was reduced by 19,31% when the subject walked with the exoskeleton. These results indicate that the gait using the robotic exoskeleton was faster and more balanced than the gait with the passive supports.

Table 1: Kinematic outcome measures with passive supports (Exp. 1) and robotic exoskeleton (Exp. 2).

	Exp. 1: Supports	**Exp. 2: Exoskeleton**	**% 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. disp. (cm)	*7,89*	*6,37*	*–19,31*

4 CONCLUSIONS

This paper presents the mechanical design and control of a patient-tailored, low-cost and low-weight robotic exoskeleton to assist the gait of subjects with SCI. The main innovation lies in its modular design that allows to install the technology to the current passive orthopaedic supports that are owned by the patient. This allows to continue rehabilitation outside the clinical environment more easily, due to its affordable price and lightness. The modular components of the exoskeleton are a compact knee actuation system, composed by an electrical motor and a Harmonic Drive gearbox; an inertial measurement unit at the shank to detect user intention; and a backpack worn by the subject. The backpack makes it a portable device and contains a BeagleBone Black board, the motor drivers and the battery.

This work reports a preliminary experimental assessment of the presented assistive device on a female subject with incomplete SCI. Three-dimensional kinematic motion analysis shows that the subject walked faster, and in a more balanced and stable way when wearing the robotic exoskeleton as compared to the case when the subject used her passive supports. 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 exoskeleton.

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.

Acknowledgements

This work has been supported by the Spanish Ministry of Economy and Competitiveness under the project DPI2015-65959-C3-2-R, co-funded by the European Union through ERDF funds.

References

[1] J. Beil, G. Perner, T. Asfour. Design and control of the lower limb exoskeleton KIT-EXO-1. IEEE International Conference on Rehabilitation Robotics, Singapore, (2015).

[2] J. Bickenbach, et al. International Perspectives on Spinal Cord Injury, World Health Organization (WHO), (2013). ISBN: 978924156466 3.

[3] J. Blaya, H. Herr. Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait, IEEE Transactions on Neural Systems and Rehabilitation Engineering, 12, (2004), 24-31.

[4] M. Bortole, A. Venkatakrishnan, F. Zhu, J.C. Moreno, G.E. Francisco, J.L. Pons, J.L. Contreras-Vidal. The H2 robotic exoskeleton for gait rehabilitation after stroke: early findings from a clinical study, Journal of Neuroengineering and Rehabilitation, 12, (2015), 1-14.

[5] G. Bovi, M. Rabuffetti, P. Mazzoleni, M. Ferrarin. A multiple-task gait analysis approach: kinematic, kinetic and EMG reference data for healthy young and adult subjects, Gait and Posture, 33, (2011), 6-13.

[6] G. Colombo, M. Wirz, V. Dietz. Driven gait orthosis for improvement of locomotor training in paraplegic patients, Spinal Cord, 39, (2001), 252-255.

[7] P.L. Ditunno, M. Patrick, M. Stineman, J.F. Ditunno. Who wants to walk? Preferences for recovery after SCI: a longitudinal and cross-sectional study, Spinal Cord, 46, (2008), 500-506.

[8] A.M. Dollar, H. Herr. Lower extremity exoskeletons and active orthoses: challenges and state-of-the-art, IEEE Transactions on Robotics, 24, (2008), 1-15.

[9] A. Esquenazi, M. Talaty, A. Packel, M. Saulino. The ReWalk powered exoskeleton to restore ambulatory function to individuals with thoracic-level motor-complete spinal cord injury, American Journal of Physical Medicine and Rehabilitation, 91, (2012), 911-921.

[10] J.M. Font-Llagunes, R. Pàmies-Vilà, J. Alonso, U. Lugrís. Simulation and design of an active orthosis for an incomplete spinal cord injured subject, Procedia IUTAM, 2, (2011), 68-81.

[11] J.M. Font-Llagunes, U. Lugrís, F. Romero, D. Clos, F.J. Alonso, J. Cuadrado. Design of a patient-tailored active knee-ankle-foot orthosis to assist the gait of spinal cord injured subjects. International Workshop on Wearable Robotics, Baiona, Spain, (2014).

[12] C. Hartigan, C. Kandilakis, S. Dalley, M. Clausen, E. Wilson, S. Morrison, S. Etheridge, R. Farris. Mobility outcomes following five training sessions with a powered exoskeleton, Topics in Spinal Cord Injury Rehabilitation, 21, (2015), 93-99.

[13] S. Hesse, D. Uhlenbrock. A mechanized gait trainer for restoration of gait, Journal of Rehabilitation Research and Development, 37, (2000), 701-708.

[14] M. Hubli, V. Dietz. The physiological basis of neurorehabilitation – locomotor training after spinal cord injury, Journal of Neuroengineering and Rehabilitation, 10, (2013), 1-8.

[15] H. Schmidt, C. Werner, R. Bernhardt, S. Hesse, J. Krüger. Gait rehabilitation machines based on programmable footplates, Journal of Neuroengineering and Rehabilitation, 4, (2007), 1-7.

[16] E. Strickland. Good-bye, wheelchair, IEEE Spectrum, 49, (2012), 30-32.

[17] A. Tsukahara, Y. Hasegawa, K. Eguchi, Y. Sankai. Restoration of gait for spinal cord injury patients using HAL with intention estimator for preferable swing speed, IEEE Transactions on neural systems and rehabilitation engineering, 23, (2015), 308-318.

[18] J.F. Veneman, R. Kruidhof, E.E.G. Hekman, R. Ekkelenkamp, E.H.F. Van Asseldonk, H. Van der Kooij. Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation, IEEE Transactions on Neural Systems and Rehabilitation Engineering, 15, (2007), 379-386.

[19] T. Yakimovich, E.D. Lemaire, J. Kofman. Engineering desing review of stance-control knee-ankle-foot orthoses, Journal of Rehabilitation Research and Development, 46, (2009), 257-267.