# A Powered Lower Limb Orthosis to Assist the Gait of Incomplete Spinal Cord Injured Patients

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## Abstract

*This paper addresses the mechanical design and control of a new active stance-control knee-ankle-foot orthosis. The orthosis is intended to provide gait assistance for incomplete spinal cord injured patients with functional hip muscles, but partially denervated knee and ankle muscles. This device consists of a passive compliant joint that constrains ankle plantar flexion, along with a powered knee unit that prevents knee flexion during stance and controls flexion-extension during swing. For this purpose, the knee joint incorporates a controllable mechanical locking system and an electrical DC motor. Based on human walking biomechanics, a hybrid control model is proposed. This model takes into account the parameters of the orthosis and the characteristics of the gait cycle, which is divided in eight different phases. A fractional order controller is designed following decision based control techniques.*

## 1. Introduction

Spinal cord injuries (SCI) cause paralysis of the lower limbs as they break the connections from the central nervous system to the muscular units of the lower body. There are different SCI levels according to the standard neurological classification of SCI of the American Spinal Injury Association (ASIA). Those are classified by the ASIA Impairment Scale (AIS) and range from A (complete SCI) to E (normal motor and sensory function).

The developed active orthosis is aimed at assisting incomplete SCI subjects with AIS level C or D. In these cases, the motor function is preserved below the neurological level (lowest segment where motor and sensory functions are normal), being the difference between C and D the muscle activity grade of the key muscular groups. The target patients can perform a low-speed, high-cost pathological gait by using walking aids such as crutches, canes or parallel bars. The energy cost and aesthetics of this walk can be improved by means of active orthoses [1,2], which require external actuation mechanisms to control the motion of the leg joints during the different phases of human gait. These devices may also be useful to achieve some degree of neuro-rehabilitation and recover lower limb muscle control.

Orthoses can be classified according to the joint for which they are designed. The function of an ankle-foot-orthosis (AFO) is to guide ankle plantar and dorsiflexion. In some cases, such as cerebral palsy or SCI, the AFO is used to avoid excessive plantar flexion which is one of the causes of drop-foot gait [3]. The knee-ankle-foot-orthosis (KAFO) is used by patients with more severe gait dysfunctions, including partial or complete paralysis of the lower limbs [4]. One particular type of KAFO is the stance-control knee-ankle-foot-orthosis (SCKAFO), which is well-suited for patients with incomplete SCI [1,5]. This device permits free knee motion during swing and locks knee flexion during stance. There are also hip-knee-ankle-foot-orthoses (HKAFO) that assist all lower limb joints.

These orthoses can be passive mechanisms that support weakened or paralyzed body segments, or active devices that assist joint motion using an external power source. Blaya and Herr [6] developed and active AFO to assist drop-foot gait. This is based on a linear series elastic actuator to assist ankle motion and it uses plantar sensors and potentiometers to detect the gait phases. Quintero *et al.* [7] presented a powered lower limb orthosis that controls hip and knee motion with brushless DC motors. This device is used together with a standard AFO and includes joint potentiometers and accelerometers located at the thigh for control purposes. Kawamoto and Sankai [8] developed the HAL (Hybrid Assistive Limb), an exoskeleton actuated by rotary motors, which is designed to assist the lower limbs of elderly people. Its control system generates external joint torques based on EMG measurements, foot-ground contact forces and joint angles. Other commercial devices aimed at assisting the gait of paraplegic individuals are the ReWalk (Argo Medical Technologies Ltd.) and the eLEGS (Berkeley Bionics). Pneumatic artificial muscles (McKibben muscles) are also used in orthotics as in [9], where a KAFO consisting of 6 artificial muscles that mimic the agonist-antagonist pairs of the human body is presented.

In this paper, we present a new design of an active SCKAFO aimed at assisting incomplete SCI patients that preserve normal motor function of the hip flexors and extensors, but have partially denervated muscles at the knee and ankle joints. The detailed mechanical design of the developed assistive device is presented. The knee unit actuates the knee flexion-extension during swing and locks the knee flexion during stance by means of a commercial controllable locking system. The selection of the knee motor is based on the simulation of the combined human-orthosis actuation using an optimization approach previously presented in [10]. The ankle unit includes a passive joint that applies a torque so as to avoid drop-foot gait and limits ankle dorsiflexion. Both units are modular and are easily adapted to standard orthoses. The proposed device is equipped with sensors for its autonomous operation. To control the orthosis, it is necessary to identify the gait cycle and establish locking and actuation phases. The design of the control will define the locomotor adaptation as exposed in [11]. In this work, four plantar sensors on each foot sole will be used to identify the different gait phases and then, a reference will be defined (for each phase) to control the orthosis.

## 2. Biomechanical specifications

The proposed active SCKAFO should control the flexion-extension motion of the knee joint during the swing phase and lock the knee flexion at any knee angle during the stance phase. According to the orthopedists, the actuation of the ankle motion is not essential to allow SCI subjects with AIS C and D levels to walk. The functions needed at this joint are passive dorsiflexion assistance during initial stance, to compensate for contact forces at heel landing, and during swing to avoid drop-foot (or equinus) gait.

This device has to include several sensors (in both the knee and ankle units) for its autonomous control. The orthosis must include plantar sensors on the insoles to detect foot-ground contact and also the period within the stance phase (initial contact, mid-stance, terminal stance). Angular sensors at both joints are also needed to detect the gait phase and they are also used as inputs of the actuation control system. Besides the previous biomechanical and control specifications, the desired orthosis should be lightweight (a weight lower than 2.5 kg per leg is desirable), be quiet in operation, be energetically efficient, be low cost, and support a large segment of potential users.

## 3. Mechanical design of the active orthosis

The ankle unit is based on a commercial passive AFO which has been minimally modified in order to adapt an optical encoder that measures the ankle joint angle. This is composed of two supporting aluminum bars which are adjusted to the shank by means of hook-and-loop straps. In both sides of the ankle, there is a passive *klenzak* joint that applies the needed passive torque. The ankle unit constrains the dorsiflexion angle of the ankle to be between 0 and 20º, thus avoiding drop-foot gait. As it has been mentioned, an optical encoder (with a resolution of 4096 counts per turn) has been placed on this joint to measure the ankle angle. Each ankle unit is also equipped with a set of four contact sensors which are placed under the orthosis insole and they allow to know if contact is on the heel (initial contact), on the mid-foot (mid stance) or on the toes (terminal stance).

The solution adopted for the design of the knee unit consists of two independent systems to assist swing and stance. The swing flexion-extension motion is controlled by means of an electrical DC motor, and a commercial electronically controllable locking mechanism is used to prevent knee flexion during stance. The design of the knee unit is shown in Figure 1. The motor and the locking system are placed at the lateral and medial sides of the knee, respectively.

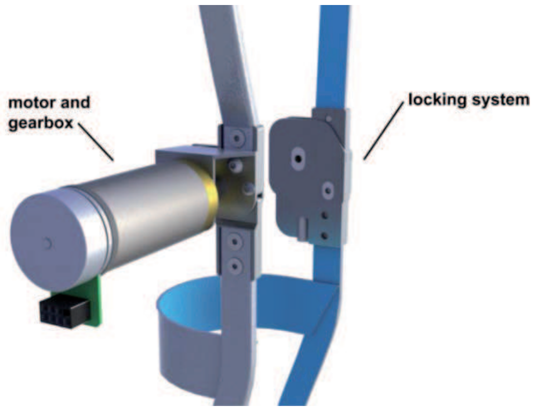


***Figure 1.*** *Knee unit of the SCKAFO.*

The commercial locking mechanism locks knee flexion by means of a pre-loaded spring that pushes a locking pawl into a toothed ring. This locking is passive and does not require energy consumption compared to other locking devices, like electrical brakes or clutches. To unlock the joint, a solenoid moves a plunger against the spring tension and the locking pawl falls out of the toothed ring. It is important to remark that in case of power or control failure, this system stays mechanically locked which is the safest situation. In this case, the patient can stand (with the knees fully extended) and walk with the aid of crutches, canes or parallel bars.

In order to select the proper knee motor, an optimization approach to quantify the simultaneous contributions of muscles and orthosis to the net joint torques of the human-orthosis system was used [10]. In this approach, the net joint torques are first obtained through inverse dynamics using a multibody biomechanical model and kinematic data. Then, the muscle-orthosis redundant actuator problem is solved through a physiological optimization approach with a cost function that accounts for muscular and robotic energy consumption. In this approach, muscles are modeled as Hill-type actuators, and a weakness factor is used to limit muscle activation in partially functional muscles. According to the simulations, the maximum orthosis torque needed during swing is 11.5 Nm, and the partially functional muscles that control the knee motion provide 8.5 Nm.

The selected motor is a "Maxon Motor EC45 flat" with a planetary gearbox (reduction 1:156). The motor produces a maximum torque of 15 Nm and a maximum angular velocity of 43 $min^{-1}$. The weight of the motor including the gearbox is 0.58 kg. The actuator is equipped with an optical encoder with a resolution of 500 counts per turn (of the motor). The encoder data serve to calculate the joint angle and angular velocity which are used as inputs of the actuation control system.

The global design of the orthosis is shown in Figure 2. This orthosis is lightweight, the total weight of the orthosis is about 1.9 kg (motor: 0.58 kg, locking system: 0.40 kg; lateral bars and other components: 0.92 kg), and it is also energetically efficient. Moreover, the design may support a large segment of potential users, since it is adaptable to different subjects and different levels of dysfunction. This design represents a first prototype of the active orthosis that will be used in a lab environment to study the combined human-orthosis actuation and adaptation. The required sensors, the actuator and the locking system will be powered by means of an external supply unit.


***Figure 2.*** *Global mechanical design of the orthosis.*

## 4. Control approach

To achieve the functional specifications of the active orthosis during the gait cycle, control tasks are necessary. In order to identify the current gait cycle phase to assist the control, eight major events are identified by four plantar sensors (MotionLab Event Switch MA-153) on each foot. As shown in Figure 3, the output of the plantar sensors together with left and right knee angles ($\theta_{kl}$ and $\theta_{kr}$) are considered as a feedback to the controller. The goal is to introduce the proper input signal to track the reference angle in each phase. The decision based controller will recognize which phase of the gait cycle is active and send the proper reference to the controller: assistance for the movement during the swing phase and locking during the stance phase.

Figure 4 shows the decision based controller, which will send a command to lock the motor during stance phase (keep the angle constant) and to follow the calculated reference during swing phase. In order to design a controller based on these specifications, a hybrid model is used, which takes into account the parameters of the DC motor and the orthosis and also the characteristics of the gait cycle by means of the corresponding joint angles (Figure 5).

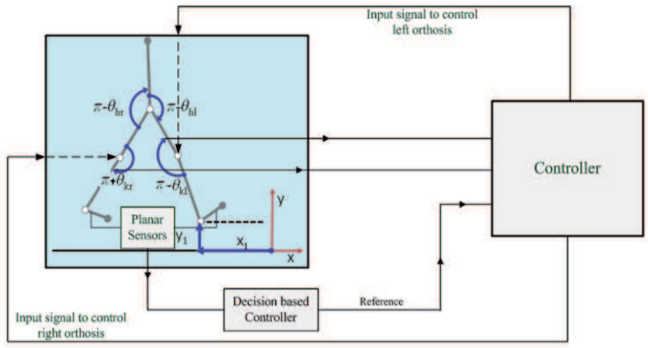


***Figure 3.*** *Block diagram of the controlled system.*

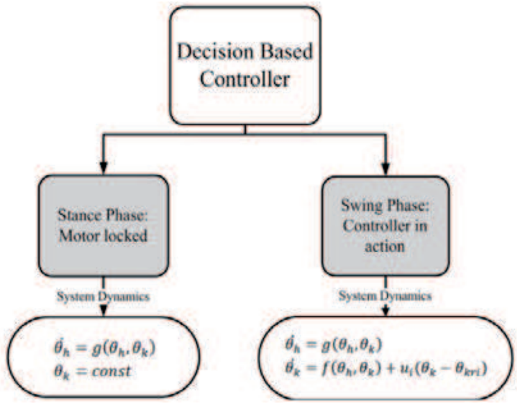


***Figure 4.*** *Decision based controller.*

Using the parameters of the DC motor and the orthosis, the model can be easily represented by:

$$P(s) = \frac{\theta_0}{V_{in}} = \frac{3.58}{s(0.01s+1)} \qquad (1)$$

where $\theta_0$ and $V_{in}$ are the motor output angle and the input voltage, respectively. A fractional order PI controller (FPI) can be considered as:

$$C(s) = K_p + K_i s^{-\lambda},\ 0 < \lambda \le 1 \qquad (2)$$

In order to tune the parameters $K_i$ and $K_p$, the following considerations will be taken into account [12]:

I) Phase margin specification

$$Arg\left(C(j\omega_c)P(j\omega_c)\right) = -\pi + \phi_m$$

II) Robustness to variation in the gain of the plant

$$d\frac{Arg\left(C(j\omega)P(j\omega)\right)}{d\omega}\Bigg|_{\omega=\omega_c} = 0$$

III) Gain crossover frequency specification

$$\left|C(j\omega_c)P(j\omega_c)\right| = 1$$

By means of those specifications, the controller parameters $K_i$ and $K_p$ can be obtained as:

$$K_p = \frac{\omega\sqrt{1+(\omega T)^2}}{G\sqrt{\left(1+K_{ip}\omega^{-\lambda}\cos\frac{\lambda\pi}{2}\right)^2 + \left(K_{ip}\omega^{-\lambda}\sin\frac{\lambda\pi}{2}\right)^2}} \qquad (3)$$

$$K_i = K_{ip} K_p \tag{4}$$

with

$$K_{ip} = \frac{-\tan\left(\phi_m + \tan^{-1}(\omega_c T) - \frac{\pi}{2}\right)}{\omega^{-\lambda}\left(\sin\frac{\lambda\pi}{2} + \cos\frac{\lambda\pi}{2}\tan\left(\phi_m + \tan^{-1}(\omega_c T) - \frac{\pi}{2}\right)\right)} \tag{5}$$

The gain crossover frequency is set as $\omega_c$ = 100 rad/s and the desired phase margin as $\phi_m$ = 40º. For the $PI^{\lambda}$ controller, the parameters obtained are: $K_i$ = 86.87, $K_p$ = 37.48 and $\lambda$ = 0.68. Mean square error (MSE) of this method $MSE(PI^{\lambda})=0.0025$ imply a quick response of the controller (see simulation results in Figure 6). It must be mentioned that, in the simulations, the controller is applied in both the stance and swing phases of gait in order to show its performance. However, in practice, the controller will only be activated during the swing phase and the knee joint will be locked during stance phase.

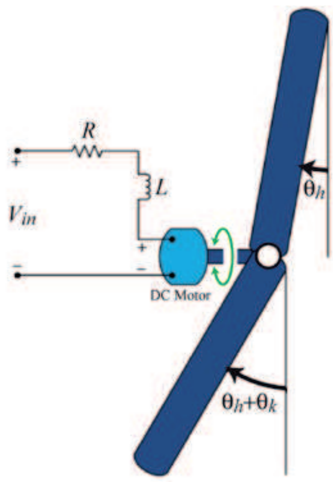


***Figure 5.*** *Mechanical orthosis model.*

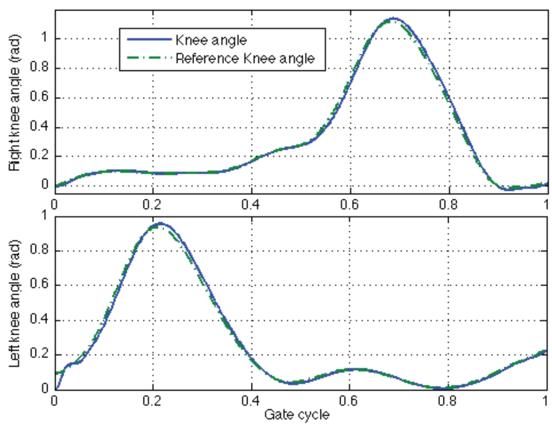


***Figure 6.*** *Simulation result of the controlled knee joint.*

## 5. Conclusions

The mechanical design of an active stance-control knee-ankle-foot-orthosis is presented. The presented orthosis is aimed at assisting incomplete SCI subjects with partial denervation at the knee and ankle muscles. The ankle unit includes a passive *klenzak* joint, and the knee unit is composed of a mechanical locking system that prevents knee flexion during stance and an electrical actuator that assists knee flexion-extension during swing. The prototype is equipped with the required sensors for its autonomous operation (encoders at joints and plantar sensors). The main advantages are its light weight and modularity. Regarding to control tasks, a fractional order controller is proposed. The controller parameters are tuned using proper specifications based on human gait cycle. This controller is not sensitive to the noise and eliminates the steady state error. The simulation results show the efficiency of the controller.


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## References


[1] Yakimovich T, Lemaire ED, Kofman J. Engineering desing review of stance-control knee-ankle-foot orthoses. *J Rehab. Research & Development*, vol. 46, 2009, pp 257-267.

[2] Dollar AM, Herr H. Lower extremity exoskeletons and active orthoses: challenges and state-of-the-art. *IEEE T Robotics*, vol. 24, 2008, pp 1-15.

[3] Romkes J, Brunner R. Comparison of a dynamic and a hinged ankle–foot orthosis by gait analysis in patients with hemiplegic cerebral palsy. *Gait & Posture*, vol. 15, 2002, pp 18-24.

[4] Kaufman K, Irby S, Mathewson J, Wirta R, Sutherland D. Energy-efficient knee-ankle-foot orthosis: A case study. *J Prosthetiscs & Othotics*, vol. 8, 1996, pp 79-85.

[5] Yakimovich T, Lemaire E, Kofman J, Preliminary kinematic evaluation of a new stance-control knee–ankle–foot orthosis. *Clinical Biomechanics*, vol. 21, 2006, pp 1081-1089.

[6] Blaya J, Herr H. Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait. *IEEE T Neural Systems & Rehab. Engineering*, vol. 12, 2004, pp 24-31.

[7] Quintero HA, Farris RJ, Goldfarb, M. Control and Implementation of a Powered Lower Limb Orthosis to Aid Walking in Paraplegic Individuals. *Proc. IEEE Int. Conf. on Rehab. Robotics (ICORR'11)*, Zurich, 2011, pp 953-958.

[8] Kawamoto H, Sankai Y. Power assist system HAL-3 for gait disorder person. *Lec. Notes Comp. Science*, vol. 2398, 2002, pp 19-29.

[9] Sawicki G, Ferris D. A pneumatically powered knee-ankle-foot orthosis (KAFO) with myoelectric activation and inhibition. *J NeuroEngineering & Rehabilitation*, vol. 6, 2009, pp 6-23.

[10] Alonso J, Romero F, Pàmies-Vilà R, Lugrís U, Font-Llagunes JM. A simple approach to estimate muscle forces and orthosis actuation in powered assisted walking of spinal cord-injured subjects. *Proc. EUROMECH Coll. 511 Biomechanics of Human Motion*, Ponta Delgada, 2011.

[11] Cain SM, Gordon KE, Ferris D. Locomotor adaptation to a powered ankle-foot orthosis depends on control method. *J NeuroEngineering & Rehabilitation*, vol. 4, 2007.

[12] Monje, CA. Design methods of fractional order controllers for industrial applications. Ph.D Thesis. University of Extremadura, 2006.