A powered lower limb orthosis for gait assistance in incomplete spinal cord injured subjects

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ABSTRACT
The paper deals with the mechanical design of a new active stance-control knee-ankle-foot orthosis (SCKAFO). The orthosis is intended to provide gait assistance for incomplete spinal cord injured patients that present functional hip muscles, but partially denervated knee and ankle muscles. It 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 that actuate independently. The prototype is equipped with different sensors (plantar sensors and angular encoders) for control purposes. They are used to identify the main events defining the gait phases and to provide feedback measurements for the motor control system.

Categories and Subject Descriptors
I.2.9 [**Computing Methodologies**]: Robotics – *Kinematics and dynamics, Sensors.*

General Terms
Measurement, Design.

Keywords
Active Orthosis, Human Gait, Spinal Cord Injury.

1. INTRODUCTION
The design of powered exoskeletons and orthoses that assist the human body movement is an open field of research that covers the areas of biomechanics, robotics and neurorehabilitation. Those devices are thought to help either elderly people or patients with disabilities cased by spinal cord injuries, acquired brain damage, cerebral palsy or post-polio syndrome. In this paper, we present a powered lower limb orthosis for incomplete spinal cord injured patients.

Spinal cord injuries (SCI) cause paralysis of the lower limbs as they break the connections from the central nervous system to the lower body muscles. 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]. In this paper, a biomechanical model based on multibody dynamics techniques was used and the muscles were modeled as Hill-type actuators. 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, namely, plantar sensors and angular encoders at the ankle and knee joints.

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 gait. Figure 1 shows the human gait cycle and the specified functions of the orthosis at the knee and ankle joints.

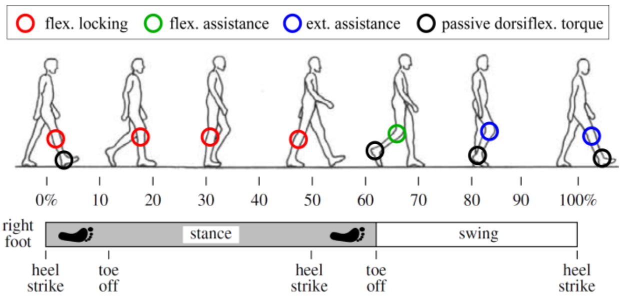

Figure 1. Functional specifications of the active orthosis during the gait cycle.

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. These sensors will be used to switch between the stance and swing modes, this has to be done with a reaction time below 6 ms [1]. 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

This section presents the mechanical design of the ankle and knee units of the presented active orthosis, as well as a few notes on its operation.

3.1 Design of the ankle unit

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). Figure 2 shows the CAD design of this unit.

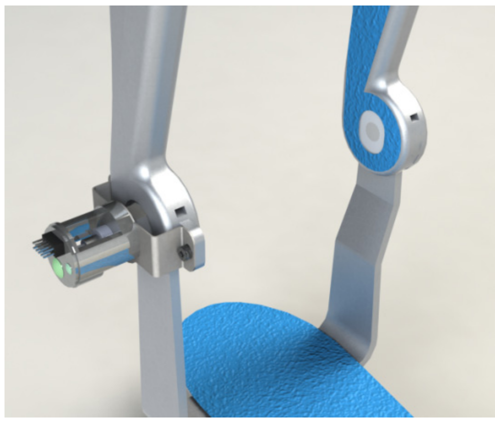

Figure 2. Detail of the ankle unit of the SCKAFO.

3.2 Design of the knee unit

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 3. The motor and the locking system are placed at the lateral and medial sides of the knee, respectively.

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.

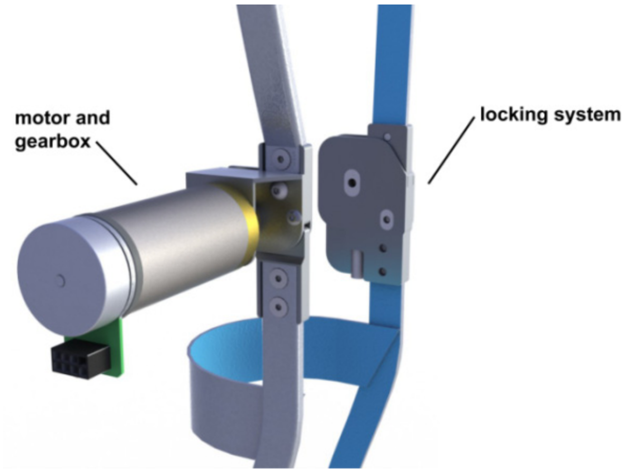

Figure 3. Knee unit of the SCKAFO.

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 were first obtained through inverse dynamics using a multibody biomechanical model and kinematic data. Then, the muscle-orthosis redundant actuator problem was solved through a physiological optimization approach with a cost function that accounts for muscular and robotic energy consumption. In this approach, muscles were modeled as Hill-type actuators, and a weakness factor was used to limit muscle activation in partially functional muscles. According to the simulations, the maximum orthosis torque that is 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 specifications of the motor plus the gearbox are shown in Table 1.

Table 1. Specifications of the motor and gearbox group at the knee unit.

Specification	Value
Nominal/Maximum torque	10.88/15 Nm
Nominal/Maximum angular velocity	33.65/43 min^{-1}
Nominal voltage	18 V
Nominal current	3.54 A
Weight/Length	0.58 kg/87.8 mm

According to the results in [10], the actuator provides enough torque to assist the musculoskeletal system of the SCI patient during the swing phase. The actuator is equipped with an optical encoder with a resolution of 500 counts per turn (of the motor), which results in 78000 counts per turn of the gearbox axis. The encoder data serve to calculate the joint angle and angular velocity which are used as inputs of the actuation control system. It must be pointed out that linear actuators were also considered at the initial design stage. However, this option was rejected because adequate actuators in terms of volume and weight did not fulfill the required velocity specification.

3.3 Global design and orthosis operation

The global design of the orthosis is shown in Figure 4. The figure on the right hand side shows the real prototype. 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 4. Global mechanical design of the SCKAFO: CAD design (left) and real prototype (right).

The operation of the orthosis during the gait cycle is as follows: At initial stance, the plantar sensors detect contact and then the knee joint is locked; during this phase the motor does not exert any torque on the joint. During the stance phase the plantar sensors and ankle encoder data give information on the evolution of the gait cycle. Once contact is over (because the other leg has landed on the ground), the solenoid of the locking mechanism turns on to unlock the knee joint. Then, the swing phase begins and the knee actuator assists the knee flexion and then extension. The motor control during this phase will be done based on the motor and ankle encoders. After the swing phase, the leg makes contact again with the ground and the new stance phase begins. At this moment, the control system is aimed at controlling the walking motion. In the future, we plan to detect and control other states such as standing or sitting down.

4. 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 main advantages are its light weight and modularity.

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). In the future, the use of passive elements, such as springs, to improve the energy efficiency of this assistive device will be explored.

5. ACKNOWLEDGMENTS

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