A Light-weight Exoskeleton for Hip Flexion-extension Assistance

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Abstract: Wearable robots can represent a way to rehabilitate and/or assist people affected by gait disorders that are common problems associated with ageing, following orthopedic surgery or diseases like brain injuries. In order to improve their usability and effectiveness, exoskeletons aiming at assisting people affected by gait impairments should be light-weight devices and have safe and low output impedance actuators. In this paper we present a light-weight exoskeleton endowed with two series elastic actuators for hip flexion-extension assistance.

1 INTRODUCTION

Gait disorders and lower-limb impairments are common and often devastating companions of ageing (Snijders, 2007). Old age is the most important risk factor for gait disturbance (Stolze, 2005): several population-based study has shown a 35% prevalence of gait disorders among persons over age 70 (Verghese, 2006), and 80% over 85 years of age.

Gait disorders lead to several major consequences, including falling (leading to major fractures or head trauma), the number of which is expected to reach 500,000 by the year 2040, leading to a total annual cost of 16 billion dollars (Burge, 2007). Another important consequence is reduced mobility, which leads to loss of independence. Gait disorders are also associated with reduced survival, which can be attributed to a combination of fatal falls, reduced cardiovascular fitness, and death from an underlying disease (Snijders, 2007; Wilson, 2002; Verghese, 2006).

Wearable robotic orthoses (namely "exoskeletons") capable to rehabilitate and/or assist people affected by gait disorders have been proposed as a solution by many research teams active in the field of rehabilitation robotics (Pons, 2010; Ronsse 2011a; Ronsse 2011b). Both unilateral and bilateral robotic exoskeletons are available in the current state of the art and relevant for this study (Dollar, 2008).

Relevant unilateral orthoses – among many – are: ALEX, a 2 DoF powered leg exoskeleton (Banala, 2006); the ankle-foot and knee-ankle-foot orthoses powered by McKibben type pneumatic muscles developed by Sawicki (Sawicki, 2005), SERKA, an active knee orthosis for gait training (Sulzer, 2009); AKROD, a knee orthosis with an electro-rheological fluid (ERF) variable damper (Weinberg, 2007); the knee-ankle foot orthosis developed at the Vrije Universiteit Brussel, driven by pleated pneumatic artificial muscles (Beyl, 2010).

Among the many bilateral orthoses we can mention: LOPES, a lower-limb exoskeleton for post-stroke rehabilitation and driven by series-elastic actuators (Veneman, 2007); ReWalk, a bilateral robotic suit for the mobility of paraplegic patients (Argo Medical Technologies, Israel); the Walking Assistance devices from Honda; HAL, a powered suit for motion assistance commercialized by Cyberdyne (Tsukuba, Japan). Relevant for this study are also the human performance augmenting exoskeletons for the lower limbs such as: BLEEX (Kazerooni, 2006), the Sarcos exoskeleton (Sarcos, US), and the quasi-passive exoskeleton developed at MIT by Walsh (Walsh, 2007).

In this paper, we introduce the design of a light-weight exoskeleton for the assistance of hip flexion-

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extension. The design of this device embeds two main innovative solutions. First, it has a novel, compact and light-weight series-elastic actuation unit. Second, carbon-fibre linkages embedding passive degrees of freedom (DoF) are used to ensure good kinematics compatibility, thus enhancing the comfort of the human-robot physical interaction. Finally, in this paper, we also present the results of the experimental characterization carried out to assess the performance of the actuation and control system.

2 METHODS

This exoskeleton is conceived for providing torque to the user's flexion-extension hip joint. It is constituted (Figure 1) by a frame carrying an actuation unit and a upper-leg link for each side. The frame is worn on the user's trunk, and is made up of an extensible rear bar coupled with two carbon-fibre lateral arms: it allows a fast don-doff procedure thanks to a detachable pin and a fine tuning of the size on medial-lateral direction thanks to a leadscrew driven adjustment. Two thermoformed orthotic shells (customized on the user's body) interface the frame with the user's trunk ensuring a comfortable wearing and avoid slippage during the application of the assistive torque. Two straps allow portion of the structure's weight to be supported by the shoulders, avoiding excessive pressure on the wearer's trunk.

The two actuation units are mounted on the frame lateral arms, and their position can be adjusted in order to align them with the hip flexion-extension human axes by means of two lockable sliders.

Figure 1: CAD model of the exoskeleton.

Coupled to the actuation axes there are two carbon-fibre linkages, molded with a shape that sweeps from the lateral to the back side of the thigh.

The shape of the links allows to swap the two, connecting the actuation axes with the front side of the thigh, without compromising the functionality. This possibility turns useful when it is necessary to maintain the rear of the thighs free from any component, for instance to allow the user to sit without hindrance.

The interface with the thighs is provided by orthotic shells tightened around the limbs by means of elastic belts. In order to fit different length of lower-limbs the vertical position of the plastic shells is adjustable thanks to lockable sliders. Thigh links are also endowed with a passive rotational DoF for abduction-adduction: this joint is located in a distal position with respect to the flexion-extension joint in order to let it completely passive and not loaded by the weight of the actuation unit; this passive axis is not collocated with the anatomical one, but still contributes to provide a comfortable interaction.

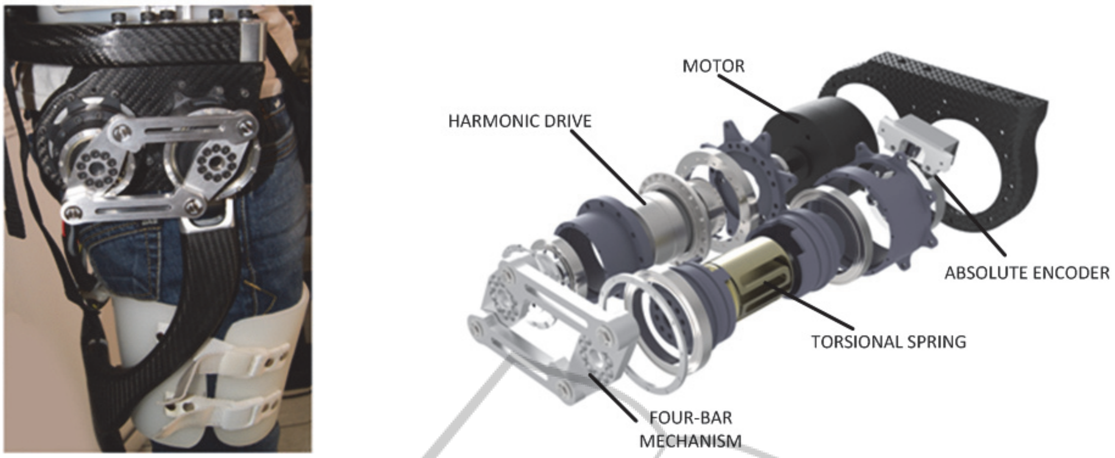
The entire system has a total weight of 4.2 kg (this weight excludes the control unit which is still remotely located in this prototype).

2.1 Actuation Units

The actuation units are two Series Elastic Actuators (SEA) (Pratt and Williamson, 1995). SEAs have been successfully applied in the field of wearable powered robots mostly to solve safety issues and reduce the inherent output impedance (Vitiello, 2013); (Veneman, 2006); (Zinn, 2004). In this case, the actuation is not rigid and allows relatively low joint impedance across the entire frequency spectrum. Furthermore, variations in the output impedance can still be achieved by means of closed-loop interaction control strategies (Pratt, 1995).

The design for the actuation units took into account hip angle and torque profiles given by Winter dataset, considering the natural cadence of 105 steps/min and a user weight of 80 kg (Winter, 2009). The target amount of assistance was set at 50% of the human torque during level walking and the maximum value of assistive torque was set at 35 N·m.

A customized torsional spring was developed to achieve a stiffness of 100 $N \cdot m \cdot rad^{-1}$, which is a value comparable with the average stiffness of the human hip during level walking (Walsh, 2007): this value prevents the subject from an uncomfortable (or even painful) interaction with an excessively stiff device in case of high frequency movements



(a) (b)

Figure 2: Lateral view (a) and exploded view (b) of the actuation unit.

(e.g. spasms, interaction with the ground). The same spring was used for the actuation of the elbow exoskeleton (NEUROExos) developed at The Biorobotics Institute (Cempini, 2013: see the reference for details on the torsional spring).

Each actuation unit (Figure 2) is configured around two parallel axes. On one axis there is a DC motor (Maxon Motor® EC60, 100W) equipped with an incremental encoder and coupled with a Harmonic Drive (Harmonic Drive® CPL-17A-080-2A). On the other axis (which is the actual hip joint axis) there is the torsional spring in series with a 32-bit absolute encoder (Renishaw® RESOLUTE™, ring: RESA30USA052B, readhead: RA32BAA052B30F), which actually measures the hip joint angle. The transmission between two parallel axes is obtained by means of a 4-bar mechanism. This configuration was chosen in order to have a small lateral encumbrance, however the swinging of the arms could be not as natural as desired. In order to overcome this limitation in the next version of the exoskeleton the actuation units will be placed on the rear part of the support frame.

2.2 Control

The control system runs on a real-time controller, a cRIO9082 (National Instruments, Austin, Texas, US), endowed with a 1.33 GHz dual-core processor running a NI real-time operating system and a Field programmable gate array (FPGA) processor Spartan-6 LX150. Motor velocity is controlled by means of a commercial servo (Maxon EPOS2 70/10). On top of the velocity control, a closed-loop 2-pole-2-zero control is used to control the joint torque. Joint torque is estimated by measuring the deformation of the torsional spring by means of the two encoders.

3 RESULTS

Initial experiments were carried out to assess the performance of the torque control, and the usability of the device under a zero-torque control mode, i.e. with the device controlled to be as transparent as possible. Experiments showed that the torque control – when a healthy subject displaces the exoskeleton joint over a frequency range of 0.3-1.5 Hz – has a parasitic stiffness ranging from 1 to 10 N·m·rad^{-1}.

Furthermore, experiments with a subject walking with the exoskeleton on the treadmill showed that the passive DoFs allow a comfortable interaction: the parasitic interaction torque at a cadence of about 0.7 cycle/s was in the range ±1 N·m (Figure 3).

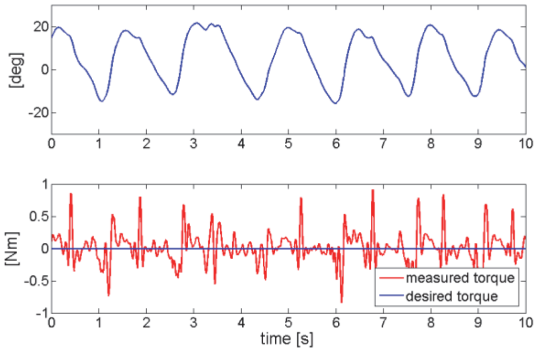


Figure 3: Joint angle profile in zero-torque control mode. *Top panel*: right hip joint angle vs. time; *down panel*: measured vs. desired torque.

Figure 4: A subject wearing the exoskeleton.

4 DISCUSSION

Recorded data and feedback from subjects (Figure 4) that tested the device showed promising performance for the developed exoskeleton and encourage to progress with a more extensive experimental characterization. The target users for the presented exoskeleton will be elderly people that need assistance to recover a more stable gait pattern; in addiction the device is addressed to reduce the metabolic consumption in lower-limb transfemoral amputees providing them with assistance during walking and other tasks. In general several kinds of patients affected by gait disorders could take advantages from the presented device.

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