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Mechanisms for actuated assistive hip orthoses

Jeremy Olivier *, Amalric Ortlieb, Mohamed Bouri, Hannes Bleuler

Laboratory of Robotic Systems (LSRO), EPFL, Lausanne, Switzerland

HIGHLIGHTS

- We presented two different hip orthoses with novel types of actuation.
- The two orthoses are optimized for assistance in various situations.
- Different characteristics are assessed in order to objectively compare our orthoses.

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ABSTRACT

Mobility is often a central problem for people having muscle weaknesses. The need for new devices to assist walking and walk related activities is therefore growing. Lower limb actuated orthoses have already proven their positive impact with paraplegic patients and are potentially promising for assisting people with weak muscles. However, the transfer from the existing systems of mobilization towards assistance implies several technical challenges as the seamless integration and the reduction of power consumption. In this paper two assistive orthoses which use different types of actuation mechanisms are presented and discussed. The first one is based on a ball screw and an excavator-like mechanism while the second one is based on a double differential actuation. Their technical capabilities are compared and contextualized for diverse activities. Objective characteristics such as the range of motion of the devices, the transparency, the maximal torque that they can provide or the RMS torque during cyclic trajectories are compared to point out which device is better adapted for specific situations.

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1. Introduction

Mobility is often a central problem for quality of life of people having muscle weaknesses. The causes of these weaknesses are diverse but lead to similar inconveniences. Difficulties to walk have consequences on physical as well as on psychological health [1]. In many cases they lead to loss of independence. With the population aging [2], the need for new technologies to assist walking and walking related activities becomes relevant.

In the last decades robots have proven their efficiency as mobilization devices [3,4]. They are able to perform repetitive tasks which are tiring for the therapists and they enable very precise control and monitoring. Motorized orthoses have even confirmed their effectiveness as mobilization devices in standing position with [5] or without [6] bodyweight support and products are currently available on the market [7]. Most of these devices were developed for spinal cord injury patients. Among other reasons, this is because such devices can successfully operate on simple position control (i.e. the device follows a predefined trajectory and imposes its motion to the user) [8].

Walking assistive devices have quite different demands on control strategies than pure mobilization robots. They need to work in interaction with their user as intuitively as possible. The person wearing an assistive orthosis decides what the movements are and when they are to be performed. The wearable robot can therefore no longer act as a mechanical admittance and impose a predefined trajectory. It must be transparent (zero impedance) when the user does not need any assistance and apply forces only when required [9]. Moreover, the mechanism should adapt to dynamic movements if the user wants to move fast. While lower limb mobilization devices often act only in the sagittal plane, the workspace of assistive device may be larger and of higher dimensionality. Thus, assistive devices are in between pure mobilization devices and exoskeletons for human augmentation.

Actuation technologies used in wearable robots are very diverse. Hydraulic, pneumatic and electrical motors have all been tested in various devices [10]. Non-standard types of actuators such as artificial muscles also seem to be promising [11]. Nevertheless, the commercial availability of all kinds of electric motors make

* Corresponding author. Tel.: +41 795403514.
E-mail address: jeremy.olivier@epfl.ch (J. Olivier).

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them the first choice for most applications. Compared to human joints, electrical motors (with a decent size) rotate fast and have a limited torque. A reduction gear train (e.g. harmonic drive) is therefore frequently used to get a significant torque [12]. Due to the relatively high transmission ratio, the low impedance required for assistive devices is difficult to achieve [9]. Some groups have therefore proposed solutions based on Series Elastic Actuators (SEA) which introduce substantial compliance between the output and the actuator [9,13]. The second drawback of a high transmission ratio is that the dynamics of the actuator is affected. The cyclic nature of walking requires that the actuator reverses direction frequently. This implies that a significant amount of torque is used to accelerate and decelerate the actuators own inertia. A possible solution to address this problem was proposed by Ryder and Sup [14] who developed a hip orthosis using a Scotch-Yoke mechanism which creates a variable transmission ratio and thus reverses automatically the direction of the actuator during walking trajectories. The electrical motors combined with well adapted transmission mechanisms seem therefore to be effective for assistive devices.

In this paper we present two distinct assistive devices which act on a single articulation: the hip. The contribution of the hip joint during walking is known to increase when people get older [15]. The hip needs to compensate for reduced strength at the knee and ankle level. The presented devices are intended to be used to study the influence of partial assistance on walking and related activities such as standing up from a sitting position or climbing and descending stairs. They are based on two different actuation mechanisms which were purposely developed for assistance and not for mobilization. The benefits and drawbacks of each solution are detailed and discussed.

2. Design specifications

To design an assistive device intended to be worn, biomechanical considerations need to be taken into account. In humans, some articulations are better approximated by ideal joints than others. The ideal model of the hip is very close to a spherical joint. We therefore consider 3 rotations around the head of the femur. The ranges of motion of these rotations are commonly assumed to be [16]:

- Extension (−)/flexion (+): $-10°$–$120°$
- Adduction (−)/abduction (+): $-30°$–$40°$
- Internal (−)/external rotation (+): $-35°$–$35°$.

As these ranges are the maximum which can be reached, they can be moderated especially when considering that the target population is people with reduced mobility. Moreover, during walking the angles are much more limited, as presented in [17].

Another important point to consider is the fact that the orthosis is placed in parallel with the joint it assists. In the case of the hip joint, the device is attached to the pelvis and to the thigh consequently creating a loop in the kinematic chain. Such loops may induce additional constraints which lock degrees of freedom (DOF). Two options are therefore conceivable to avoid reducing the mobility. Either the axes of rotation of the mechanism need to pass through the head of DOF the femur (center of the assumed spherical joint), or additional DOF are required in order to satisfy the Chebychev–Grübler–Kutzbach criterion, which states that each loop in the kinematic chain locks 6 DOF [18]. The first solution requires precise adjustments [19] and relies on skin deformations to compensate for possible misalignments. Large misalignments lead to important skin deformations and therefore to increased discomfort. The second one, while more complex (in total 6 DOF are required), guarantees that the number of DOF stays sufficient [20,21].

In order to efficiently assist the wearer, the orthotic device needs to be able to follow the wearer and to provide extra torque when required. Therefore the dynamic capabilities must at least equal the fastest movements that the wearer may perform. The maximum acceleration and velocity are estimated from typical walking trajectories and consider that the stride cadence is usually lower than 120 steps/min [22]. The peak torque a person is developing with the hip during walking is also available in [22] and is typically around 0.8 N m/kg (normalized with bodyweight). For activities such as standing up from a sitting position, a larger torque is required, especially during the initiation of the movement [23]. This torque can go up to 1 N m/kg. Even though these torques do not need to be provided entirely (especially when the movements are performed fast), their orders of magnitude are a valuable indication for the design of assistive devices. In order not to disturb the user when no assistance is required, the system must also be able to be transparent and therefore present a maximal parasitic torque of 1 N m (in zero assistance mode).

3. Presentation of the two orthoses

This section presents two distinct designs of assistive hip orthoses developed by the authors (see Fig. 1). Both devices use a 60 W motor from Maxon as their main source of power. The main difference between these two designs comes from the actuation mechanism which leads to other minor differences. Both devices have 6 DOF, one of which being actuated in order to assist the movement in the sagittal plane. This high level of mobility enables the mechanisms to preserve high ranges of motion in the three DOF of the hip. In both designs, two pivot joints are located at the pelvis level while the other four DOF are located at the joining with the thigh. The latters are composed by one prismatic joint and one spherical joint.

3.1. Ball screw driven orthosis

The ball screw driven orthosis (BSO) is based on a spindle drive mechanism (ball screw) to amplify the torque of the motor [24]. Fig. 1(a) represents the device, its kinematics and its amplification mechanism.

3.1.1. Kinematics

The first joint in the kinematic chain has its axis of rotation orthogonal to the frontal plane (see Fig. 1(a) axis number 1). The second joint is actuated and it is perpendicular to the first joint (see Fig. 1(a) axis number 2). Being second in the kinematic chain, it rotates with the first joint. This joint enables the flexion/extension movement. Due to the size of the actuation mechanism, it had to be placed second in the kinematic chain. The internal/external rotation of the leg is achievable as the last joint in the orthosis kinematic chain is a spherical joint (see Fig. 1(a)). With this kinematics, the actuation mechanism remains parallel with the leg and does not need to move to enable this movement (at least in first approximation and in the considered range of motion). However, as the hip and the spherical joint are not perfectly aligned with the leg, the first pivot joint also needs to marginally rotate. The adduction/abduction movement is more problematic. When the flexion angle equals zero, the first joint corresponds to adduction/abduction and this movement is therefore unrestricted. Conversely, when the flexion angle is 90°, the first joint is aligned with the internal/external axis of rotation of the leg thus creating a singularity. Indeed two joints enable the same movement which leads to a possible undesired rotation of the mechanism around the leg. Moreover, with this flexion configuration the adduction/abduction is hindered as no axis in the mechanism is positioned properly to enable it. A combination of large flexion and adduction/abduction is consequently impossible with this kinematics. In order to avoid

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