A Double-differential Actuation for an Assistive Hip Orthosis
Specificities and Implementation

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Abstract: The population ageing implies an increasing need for support especially in terms of mobility. Actuated orthoses offer new possibilities to assist walking by compensating the diminished muscular force which occurs with age. In order to assist efficiently the user, the orthotic device needs to provide torque without constraining the voluntary movements. Transparency is therefore a critical characteristic. A first implementation of such a device using a conventional actuation is presented and its limitations are analyzed. The walking trajectory being a cyclic movement, the actuator often needs to accelerate and decelerate. Its dynamics is therefore crucial and can be problematic at the higher cadences. Dual-differential actuation is therefore presented as a profitable alternative to overcome these weaknesses.

1 INTRODUCTION

Mobility is often a central problem for elderly people. The consequences of having difficulties to walk have an impact on both physical health and psychological well-being. With the population ageing, the need for walk assistive devices becomes therefore a central question.

Various exoskeletons have been developed for different walking assistance and rehabilitation applications (Herr, 2009). Devices such as the Lokomat (Jezernik et al., 2003) or the WalkTrainer (Bouri et al., 2006); (Stauffer et al., 2009) have demonstrated their value in particular with spinal cord injured patients. Their main characteristic is that they mobilize the wearer's leg in order to reproduce a walking trajectory. These exoskeletons therefore mainly act as admittances.

Unlike mobilization devices which impose a movement to a user who is not able to move by himself, an assistive orthosis needs to work in collaboration with the user. To enable the wearer to lead the movement, the orthosis needs to act as an impedance (Vallery et al., 2008). In the extreme case if the assistance rate tends to zero, the device needs to be fully transparent. As a consequence, it is required that the actuation mechanism is back-drivable and ideally entirely dynamically compensated.

The mechanism also needs to be dynamic enough to be able to follow the movement of the users in any situations. Walking being an cyclic movement, the orthotic device needs to be able to accelerate and decelerate accordingly to the user's motion (Ryder and Sup, 2013).

In order to be as light and as less intrusive as possible, we propose to develop devices to study the influence of single joint assistance. Therefore, this paper presents two different mechanisms to assist the movement of the hip.

2 METHODS

In this paper we describe two variants of assistive hip orthoses which were developed in the Laboratory of Robotic Systems (LSRO). The second one was developed to overcome limitations of the first variant.

The first variant is presented in section 3. The biomechanical considerations are explained and the design is described. The back drivable actuation based on a ball-screw is detailed and the limitations due to this transmission are presented. Typical walking trajectories were used to assess the dynamic capabilities of the device.

In section 4, a concept to overcome the limitations of the first variant is presented. This

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solution based on a dual-differential actuation is explained in details and a possible implementation in an orthotic device is presented in section 5.

3 FIRST VARIANT OF THE HIP ORTHOSIS

The first variant of the hip orthosis is based on biomechanical considerations such as the required torque (depending on the activity the user is performing), the velocity of the movements or the articulation range of motion (Olivier et al., 2013).

This orthosis is designed to assist the movement in the sagittal plane without constraining the other rotations of the leg. The mechanism we implemented to achieve a large range of motion and a variable transmission ratio is inspired by excavators (see fig. 1). It uses a DC motor with a ball screw transmission.

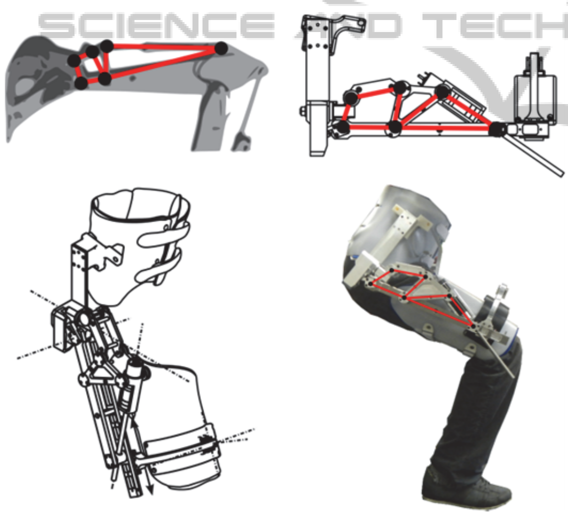

Figure 1: Amplification mechanism inspired by an excavator. This enables a large range of motion and the transmission ratio is adapted for walking as well as for standing up.

3.1 Torque and Velocity Considerations

The orthosis is aimed to assist the wearer during walking, stair climbing/descending and during the sit-to-stand transitions. The later requiring more torque especially during the first part of the movement (i.e. when the flexion angle of the hip is large), the mechanism is designed to offer a variable transmission ratio. During walking the flexion angle stays fairly small but a higher velocity is required. The smaller transmission ratio is therefore fully adapted to these requirements.

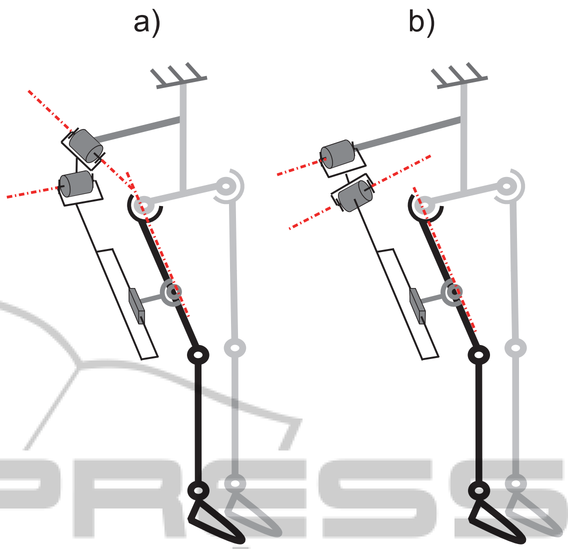

Figure 2: Kinematics of the orthosis. Six degrees of freedom are required. (a) Position of the joint in the first prototype. (b) Improved position of the rotational joints. With this configuration, the axes of rotation are always quasi-orthogonal (at least within the range of motion).

3.2 Kinematic Considerations

The hip joint can be well approximated by a spherical joint with its center being the head of the femur. Three rotations around this point are therefore considered. Aligning the mechanism’s rotations with the head of the femur being relatively complex, we decided to add three degrees of freedom (DOF) in our mechanism in order to satisfy the well-known Chebychev-Grübler-Kutzbach criterion. The mechanism being placed in parallel with the user’s hip joint, a loop in the kinematic chain is created. In order to keep the initial mobility, the mechanism’s number of DOF must be six. Two rotational DOF are therefore placed at the fixation with the pelvis and four (one translation and three rotations) are located at the thigh’s interface (see fig. 2(a)).

3.3 Performances & Limitations

The orthosis was designed to satisfy several activities’ requirements. Distinct data are then important to evaluate the performances and the limitations of the orthotic device. The maximal torque is a key value for the evaluation of the sit-to-stand transitions assistance. The velocity can be a limiting factor during dynamic movements like walking. The dynamic capabilities also determines the maximal assistance rate for walking. Eventually, the range of motion is of a major importance for the comfort in general.

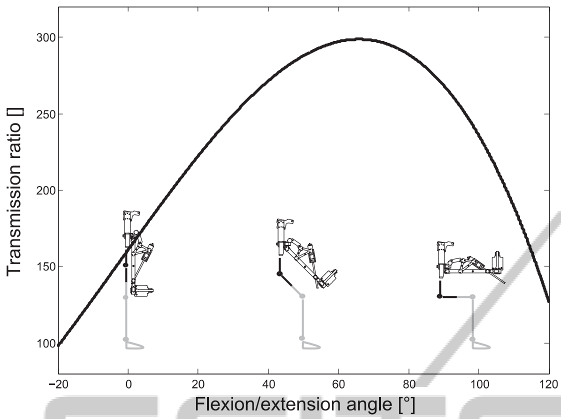

Figure 3: Transmission ratio as a function of the angle in the sagittal plane. In the walking range the transmission ratio is smaller. During standing up, the maximum torque is required when the angle is around 70°. The orthosis has is maximum transmission ratio in this range.

3.3.1 Maximal Torque and Velocity

As explained in section 3.1 a higher torque is required when the flexion angle is large (typically around 70°) for sit-to-stand transitions assistance. Fig. 3 shows the transmission ratio of the mechanism as a function of the position of the leg. During walking the peak torque is lower but the velocity is higher.

3.3.2 Assistance Rate

The assistive capabilities of the developed orthosis were evaluated by testing its dynamic performances. A typical flexion/extension trajectory was used in order to assess the required torque to make the orthosis follow the wearer during walking. The maximum assistive rate is deduced from the difference between this torque and the maximum continuous torque that the motor can provide. It was observed that the maximal assistive rate is around 30% for a 70 kg subject walking at a cadence of 100 steps/min. This rate drops to zero when the cadence increases to 120 steps/min. In that case the actuation mechanism (motor and transmission) needs all its power to accelerate and decelerate its own inertia. For more information, refer to (Olivier et al., 2013).

3.3.3 Limitation Due to the Kinematics

The amplification mechanism being relatively long, the joint enabling the flexion/extension movement had to be placed in second position in the kinematic chain (see fig. 2(a)). This configuration is sub-optimal because the rotation in the frontal plane gets locked when the flexion angle increases. Moreover, it causes a singularity when the thigh axis and the first pivot joint are aligned. This generates an internal degree of freedom − the mechanism having the possibility to rotate around the leg. This would not happen if the two rotational joints had their positions inversed (see fig. 2(b)). Indeed, the rotation in the frontal plane being limited, no singularity can be reached. In our first design the parasitic rotation is prevented by a cam mechanism.

3.4 Control

As the orthosis is intended to assist (in opposition to mobilization devices), it needs to act as an impedance. In the extreme case, the impedance is null and the device is transparent (zero assistance). To reach this very low impedance, frictional and dynamic effects are compensated. Therefore a precise model is required. Since the frictional effects are difficult to model precisely (in particular dry friction when the velocity is close to zero), the transparency is not perfect. As suggested by Zanotto et al. (Zanotto et al., 2013), force sensors placed on the supporting cuffs could be employed in order to improve transparency.

4 DUAL DIFFERENTIAL ACTUATION

In order to limit the inertia effect and the substantial induced power consumption, we propose to use a mechanism which enables to decouple the actuator from the output. As suggested by Tucker and Gassert (Tucker and Gassert, 2012), a differential mechanism could be integrated in a portable lower limb orthotic device. One of the main advantages of this kind of actuation is that the output torque can be controlled independently from the input speed of the actuator. Another advantage is that a rotational actuation can be employed which would make possible the implementation of the improved kinematics.

4.1 Clutch Principle

In order to avoid any undesired inertial or frictional effects amplified by a large transmission ratio (especially in a transparent mode), a motor/clutch mechanism can be used (see fig. 4(a)). If the motor is controlled as a velocity source, the absolute value of the output torque depends only on the clutch. Its direction however can only be in the direction of the actuator velocity.

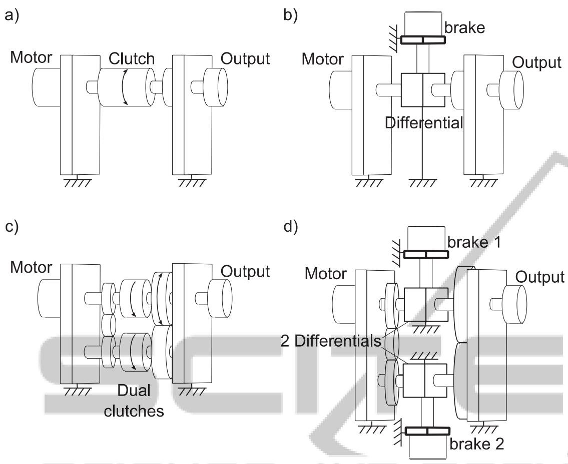

Figure 4: Clutch mechanisms. (a) Conventional clutch. The input and the output can be decoupled. (b) Differential and brake used as a clutch. (c) Dual-clutch mechanism. By means of two clutches it is possible to control the output in the two directions with an input rotating in one direction. (d) Dual-differential mechanism. It is a combination of the double clutch with the differential and brake mechanism.

4.1.1 Differential Mechanism

As mentioned by Chapuis et al., (Chapuis et al., 2007), a special case of the clutch principle can be realized with a differential and a brake (see fig. 4(b)). The main advantage of using a differential is that the transmission ratio of the motor can be adapted directly.

4.1.2 Double Clutch Principle

As presented in section 3.3.2, with a conventional actuation, a substantial amount of power is consumed to accelerate and decelerate the motor and its transmission. By using an inversion mechanism and two clutches it is possible to avoid these considerable losses. Fig. 4(c) shows the double clutch configuration. The output torque is the difference between the torques of the two clutches. Usually, if one of them is engaged, the second one should be off, in order to prevent losses. If the motor is controlled as a constant velocity source, the two clutches are used to generate the torque in both directions (Chapuis et al., 2007).

4.1.3 Dual-differential Principle

By combining the double clutch principle with the differential, a dual-differential actuator is formed. This solution was implemented by Fauteux et al. (Fauteux et al., 2010); (Fauteux et al., 2009) using a velocity source (DC motor and its reduction gear) and two magneto-rheological brakes.

4.2 Dual-differential Implementation

The differential involves a transmission ratio. By taking advantage of it, a fairly compact solution can be designed. This also enables us to have a different transmission ratio depending on the direction. This is an interesting feature in our case since, in elderly walking, the extension torque is greater than the flexion torque (JudgeRoy et al., 1996).

5 PRACTICAL REALIZATION

A differential mechanism can be realized in different ways. We implemented ours as a planetary reduction gear with two external satellites (see fig. 5(a)). The mechanism is used twice with two different transmission ratios, one of which being negative in order to be able to produce torque in both directions.

5.1 Differential based on Planetary Reduction Gear

The reduction ratio of the implemented planetary reduction gear is given by:

$$i = \frac{\theta_0}{\theta_4} = \frac{r_2 r_4}{(r_1 + r_2)(r_2 - r_3)} \tag{1}$$

where, θ_0 is the angle of the input (e.g. the motor), θ_4 is the angle of the output (e.g. the part attached to leg) and r_1, r_2, r_3, r_4 are the radii of the different gears as presented on fig. 5(a).

In a planetary reduction gear, one of the gears is fixed to the frame (gear number 1 on fig. 5(a)). In order to transform this mechanism into a differential, the gear needs to be movable and the torque applied on it will enable the control of the output torque (Fauteux et al., 2010).

An advantage of the planetary reduction gear with two external satellites is that the transmission ratio can be negative (see equation 1). Indeed if the radius r_3 is greater than r_2, the output is in the opposite direction than the input. This feature is exploited to avoid the need for an inversion mechanism on the motor side (see section 4.1.2). A schematic representation of the dual-differential based on the planetary reduction gear is presented on fig. 5(b).

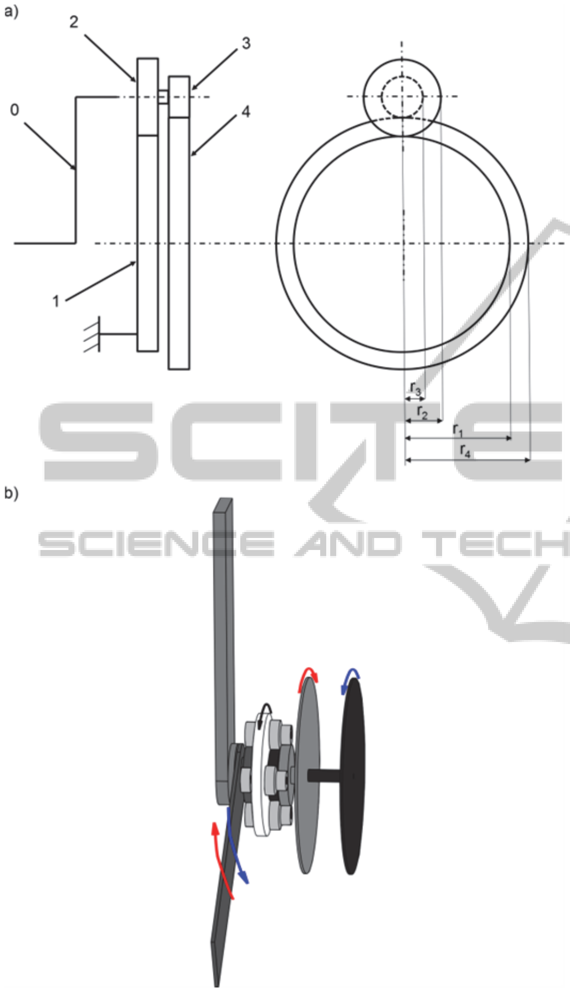

Figure 5: Differential mechanism for an orthosis. (a) The planetary reduction gear with two external satellites. In the case of a reduction gear, the gear number 1 is attached to the frame. (b) Implementation of a double differential mechanism. The motor rotates always in the same direction (black arrow). The brakes apply torques (red and blue arrows). These torques are transferred to the part attached to the leg.

5.2 Actuation

The actuation gear (velocity source) is powered by a DC motor through a worm gear. The non-back drivability is not an issue as we only use this motor as a velocity source and the output can be decoupled.

The transmission ratio between the motor and the output (calculated with the transmission ratio of the worm gear and eq. 1) is equal to 209 in one direction and to −140 in the other. The theoretical torques with a 100% efficiency are therefore respectively 18.4 Nm and −12.4 Nm for a nominal input toque of the motor. Since we are using a DC motor, this torque can be higher for a short period of time if required.

Considering that the efficiency of the transmission is around 50% mainly because of the worm gear, the RMS assistance torque is around 9 Nm. This torque typically represents around 40% of assistance for a 70 kg person walking at a cadence of 100 step/min. At the same cadence, this is about 25% more efficient than the first variant.

The brakes are cable driven bicycle disk brakes and they generate the rated output torque (there is a transmission ratio between the brakes and the output). Two 20W DC motors are used to control the torque. This solution was chosen in order to validate rapidly the concept. A more compact and reliable solution will be evaluated for the next version.

The different components of the mechanism are presented on fig. 6.

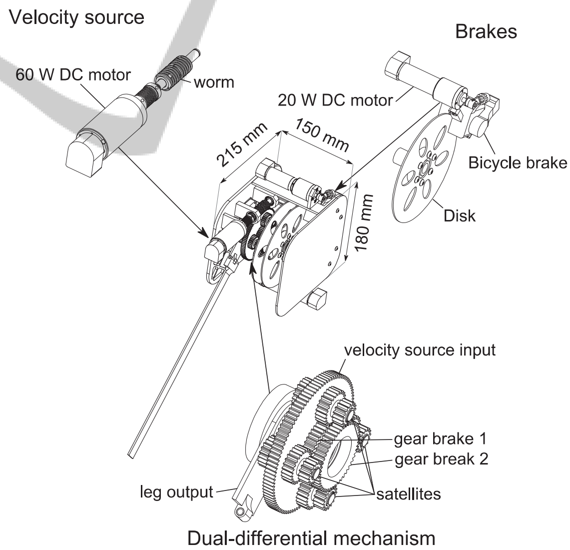

Figure 6: Hip orthosis with a double-differential actuation. The different components which constitute the mechanism are a velocity source (composed by DC motor and a worm gear), two disk brakes and the dual-differential mechanism.

5.3 Improved Kinematics

The developed mechanism makes the implementation of the kinematics presented on fig. 2(b) possible. The first rotational joint (actuated by the double-differential mechanism) corresponds to flexion/extension. The second joint enables the adduction/abduction. Since the range of motion of this rotation is limited to relatively small angles (typically 10° for adduction and 30° for abduction) no singularity can be reached. Another advantage of this kinematics is that the overall range of motion is bigger since a combination of large flexion angles with abduction or adduction is now possible.

5.4 Control

As explained in section 4, the main advantage of the dual-differential actuation is that the input (velocity source) does not have an impact on the output torque. The restriction is that the rated output velocity must be less important than the input. Under this condition, the output torque is the sum of the rated torques on the brakes (see fig. 7). Ideally the two brakes do not act simultaneously as this would unnecessarily increase the energy consumption of the system. Moreover, as the output torque depends only on the torques of the brakes, transparency is an intrinsic characteristics of the system. Indeed when the brakes are open, the output is free. In addition, only the characteristics of the brakes need to be considered to be able to precisely control the output torque.

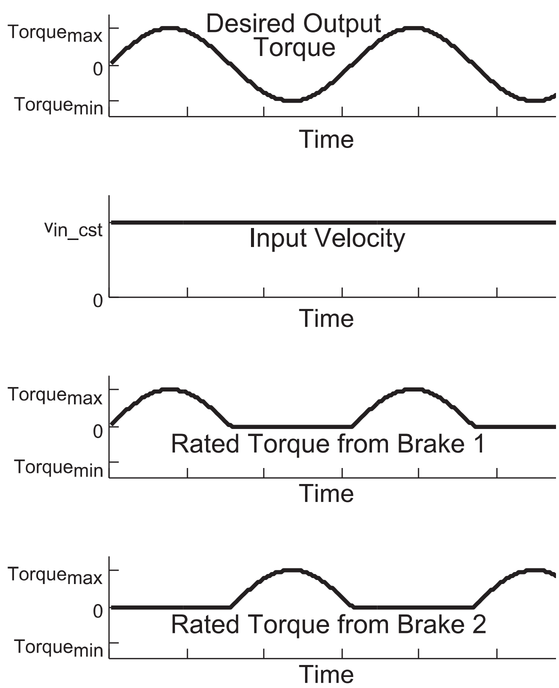

Figure 7: Graph showing the control inputs of the brakes to generate a sinusoidal torque. The velocity source does not need to adapt to the changes. In order to limit losses the brakes should not provide torque at the same time. For a positive output torque, the first brake will be engaged. The second brake will be used if a negative torque is required.

Another very important feature of the mechanism is that in case of power failure, the system becomes transparent which makes it safer than the exoskeletons with conventional actuation.

6 CONCLUSIONS

In this paper we have presented two variants of assistive hip orthoses. The first one was developed to provide a torque which is adapted to the activity the user is performing. The transmission ratio of this device is variable in order to provide an increased torque for sit-to-stand transitions and more velocity for dynamic movements like walking. Due to the inertia of the system, the assistance rate for walking depends on the cadence (i.e. the number of steps per minutes) and is therefore reduced at higher speeds. The transparency of the system is as well limited since it depends directly on the precision of the model.

The second variant uses a new type of actuation based on a novel dual-differential mechanism. It is presented as an alternative which overcomes the limitations concerning the kinematics and the rate of assistance at higher cadences. The output torque can be directly controlled by applying the corresponding rated torque on the brakes. In addition, the direction of the output torque is specified by using one brake or the other (i.e. one brake is used for flexion and the other for extension). This method has an additional intrinsic safety property as it decouples the motor automatically from the load in case of power failure. As a consequence, the system becomes transparent and the risk of accident is significantly reduced.

The two described devices are fairly powerful as we want to be able to provide a large range of assistance rate. As a consequence, their size and weight are also important (about 4 kg for one side). For later versions, a tradeoff will have to be found in order to assist efficiently the seniors while limiting the dimensions and weight of the orthosis as this could have a negative impact on their balance or on their coordination.

Further tests will be done with the two devices worn by subjects in order to validate the effects on walking or on other related activities. Both of them are useful platforms for testing different assistive strategies. The first one is very promising for testing the effects of partial assistive orthosis on sit-to-stand transitions while the second one is more adapted for dynamic and cyclic activities like walking.

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