Control of a Self-adjusting Lower Limb Exoskeleton for Knee Assistance

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Control of a Self-adjusting Lower Limb Exoskeleton for Knee Assistance

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Abstract In this paper, the authors present the control architecture of a multi-contacts lower limb exoskeleton which is designed for knee rehabilitation purposes. The device's novel mechanical architecture, which comprises passive mechanical linkages connecting the user limbs to an external rigid mechanical structure, allows a more effective control of the system's transparency as only principal torque components are transmitted to the user's anatomical joints. Different types of sensors were used to capture the user's kinematics as well as to detect contacts between the user's soles and the ground. This data allow the estimation of the gait phases in realtime, using Principal Components Analysis (PCA) and Markov chain. Finally, a predictive controller was implemented on the device to assist the knee joints during specific gait phases.

1 Introduction

Recent works on the design of exoskeletons for human anatomical joints rehabilitation have figured out that a human anatomical joint is in most cases spatial and can not be reduced to simple elementary mechanical joints (Woltring et al. 1985). Thus, the

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design of exoskeletons for functional rehabilitation must take into account the complexity of anatomical joints. A misalignment between the instantaneous anatomical axis and the mechanism's actuated axis will create residual force/torque components that may limit the natural motion of the joint, or even cause permanent injuries in extreme cases. In 2006, (Schiele and Van der Helm 2006) presented a novel design for human arm exoskeleton which has 9 d.o.f. in total. In this design, a RPR mechanism is used at the elbow joint. Only the first rotational joint is actuated while the two other joints are passive, assuring that only the actuator torque is transmitted to the elbow. It was followed by different original designs in literature such as (Cai et al. 2011; Ergin and Patoglu 2012; Stienen et al. 2009). All these authors noticed the need of using passive linkages to eliminate undesired residual force/torque components.

Several authors (Popovic et al. 1989; Rabishong et al. 1975) have conducted research on modular exoskeletons since 1970s. These systems are designed to correct an individual joint during a specific task (such as walking). However, having no contact with the ground, a modular system must have low inertia in order to not disturb the user's movement. Meantime, full lower limb exoskeletons were also developed by several research teams (Banala et al. 2006; Vukobratovic et al. 1974). They were designed to assist paraplegic patients that have full upper-body capabilities during walking. The interaction with human limbs can be measured by force/torque sensors, and the assistance is provided when necessary during walking. As all the load of the system is transferred to the ground, in theory, the device won't affect natural motions if it is able to predict and follow the user motion. By this way, designers have less concerns about creating low mass and low inertia structure as in case of modular devices. To this end, several authors have been working on the kinematic measurement of human locomotion using passive exoskeletons (Kanjanapas and Tomizuka 2013) or using camera based motion capture systems with force plates (Popovic et al. 2005). This approach seems to be very promising, especially in functional rehabilitation applications, as the device can follow the subject's motion and assist one targeted joint during specific phases.

In this paper, the authors propose a novel mechanical design for lower limb exoskeleton and a hybrid predictive control solution developed for knee assistance during locomotion. The controller has the ability to provide torque assistance to the knee joint at precise moments during the gait cycle. This can be done using a robust gait phase detection method, based on the user's joints kinematic data measured by the exoskeleton itself.

2 Mechanical Design

The authors applied the design rule propose by (Cai et al. 2011) for the determination of the total general mobility of the mechanism and the number of passive joints that should be included into the mechanism to make the whole mechanical structure free of constraints when it is attached onto an anatomical joint. The idea is quite simple: one can consider two adjacent human corporal segments as two rigid solids which

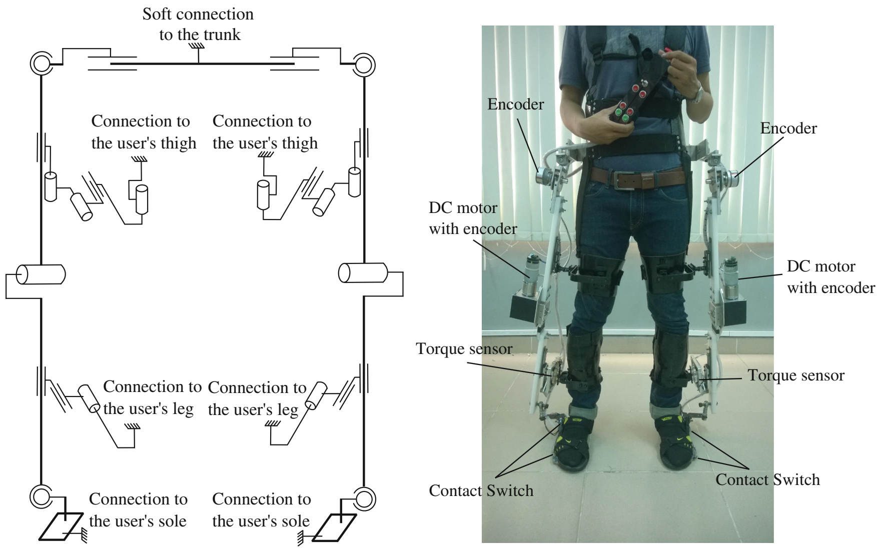


Fig. 1 The complete kinematic scheme and the first prototype of the lower limb exoskeleton

are connected to each other by an anatomical joint. In space, the relative movement of a segment relative to the other one can be described by six parameters of movement (three translations and three rotations). As an anatomical joint is composed by bones, cartilages, menisci and other flexible elements such as ligaments, tendons and so on its kinematics is always variable depending on the load condition. Therefore, one must use in total 6 kinematic parameters to describe a complete human anatomical joint kinematics. As a result, an effective modular exoskeleton must have 6 degrees of freedom (including actuated and passive mechanical joints) in order to follow the natural motion of the subject without perturbing it. Obviously, to simplify the design, passive joints performing low angular variation amplitude during the user's intended movement can be omitted. Other optimization factors such as force transmission capability or kinematic isotropy must also be considered by designers. This design rule is here generalized for full lower limb exoskeletons as shown in Fig. 1.

For the understanding of the design, one can divide the kinematic scheme into two parts: the anthropomorphic external structure which supports all the load of the system and the internal linkages, which connect the external structure to the user's corporal segments. The external structure is composed by 2 spherical joints located at the ankles, 2 actuated rotational joints located at the knees, 2 other spherical joints at the hips and 2 prismatic joints which connect the whole structure to the user's trunk. The internal linkages are composed by passive joints, which are used to connect the external structure to the human limbs (both the thighs and the legs). As it is demonstrated by several authors that the knee torque is essentially negative during walking (Kanjanapas and Tomizuka 2013; Popovic et al. 2005), the resistive motor torques can be provided at the knee during the stance phase to increase the stability of the joint or at the Preswing phase (push-off) to assist the initiation of the swing movement. During other gait phases, the two motors are controlled in such a way that the mechanism can follow the intended movement of the user. This can be done using a zero torque controller. The interactive torques measurement can be assured via the torque sensors placed between the passive mechanical linkages connecting the external structure to the human legs.

Optical encoders are used to measure the movements of the mechanical structure's principal axes of rotation, that are located at the hip joints and knee joints levels. In addition, four switches are placed at the level of the soles as well to detect contacts between the external structure and the ground during walking. With these sensors data, one can detect the user's gait phases in realtime, therefore making possible the use of a predictive control for the knee assistance during locomotion.

3 Control of the System

The human gait cycle is a time-dependent process and is often divided into two specific locomotion phases denoted by percentages which are (Perry and Burnfield 2010):

- The stance phase, from 0 % to 62 % of the gait cycle. This phase is characterized by the contact between the foot's sole and the ground.
- The swing phase, from 62 % to 100 % of the gait cycle. This phase is characterized by the sole of the foot's lack of contact with the ground.

These 2 principal phases can be decomposed into 7 sub-phases that are: Loading Response (phase 1), Midstance (phase 2), Terminal Stance (phase 3), Preswing (phase 4), Initial Swing (phase 5), Midswing (phase 6) and Terminal Swing (phase 7). For a simpler classification, the gait phases can also be assembled into 4 sub-phases that are: Heel Strike (comprising the Terminal Swing and Loading Response), Stance Phase (comprising the Midstance and Terminal Stance), Push Off (Preswing) and Swing Phase (comprising Initial Swing and Midswing). This lengthens the phases' duration, and thus improves the robustness of the detection method. The details of the gait phases classification can be seen in Fig. 2.

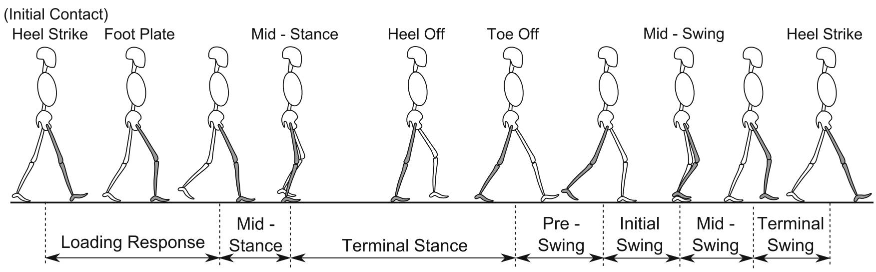
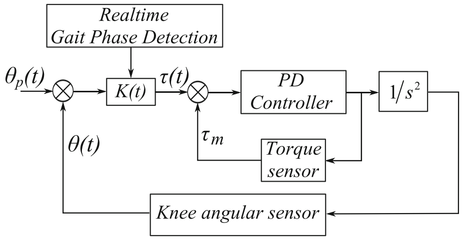


Fig. 2 Human gait phases

Fig. 3 The hybride predictive control scheme for knee assistance during locomotion



Here the switches data is coupled with the kinematic measurement of the external structure to estimate the gait phases during locomotion. Principal component analysis (PCA) is used to lower the data dimension in order to meet the requirements of estimation's accuracy and high speed computation. To enhance the reliability of the gait phases recognition, a clustering algorithm using k-means method is implemented in order to identify the mean principal components vectors for each gait phase. It is followed by the use of a Markov chain which guarantees the robustness of the recognition by removing possible detection errors issued from the first classification step, exploiting the cyclical nature of the human gait.

In the very first experiments with the device, the authors aimed to evaluate its capacity to assist the knee joint during locomotion at a precise gait phase. Here the torque assistance is provided for the knee joint exactly at the initiation of the flexion movement (i.e. at the Preswing phase) in order to increase the knee range of motion during walking activities. To this end, a hybrid predictive control architecture is proposed for the control of the interactive torques measured between the user's legs and the mechanical structure. The complete control scheme is shown if Fig. 3.

The interactive torque set-point τ is in function of the predicted knee angle $\theta_p(t)$ and the virtual stiffness $K(t)$. Its value is determined by the following equation:

$$\begin{cases} \theta_p(t) = \theta_p(t-1) + \Delta\theta(t) \\ \tau = K(t)(\theta(t) - \theta_p(t)) \end{cases} \tag{1}$$

Here $K(t)$ is an adjustable stiffness, $\theta(t)$ is the measured knee angle at time t, $\theta_p(t)$ is the predicted knee angle at time t, $\Delta\theta(t)$ is the knee angular variation at time t, which is determined in advance experimentally. Besides, $K(t)$ is in function of the gait phases. For the Preswing phase, its value can be a constant or a function of time, different from zero to create the assistance effect. Otherwise, $K(t)$ is set to zero for the other phases so that the device could follow the knee intended motion. Low-level control of the interactive torque can be realized using a simple PD or LAG regulator. The stability of the control system is assured provided that the torque sensor axis remains nearly parallel to the user's knee axis during its motion.

4 Experimental Result

Two experiments were realized. In each experiment, the user realized a 5 steps-walk with the device. The gait phase detection results as well as the measured knee angles and the interactive torques are shown in Fig. 4.

The sub-figure (1) presents the result of the first experiment. In this experiment, the motor torques were not controlled. As consequence, friction and inertia effects remained significant during the knee motion, therefore limiting its range of motion. As one can see in the sub-figure (1), the knee angular variation was only about 40° and the interactive torque amplitude was 4 $N.m$. The sub-figure (2) presents the second experiment in which the interactive torque was controlled by the predictive controller described in Sect. 3. Here one can state that the interactive torque's value increased significantly at the Preswing phases. As a result, the knee angle variation was much larger compared to that in the first experiment. Its amplitude was about 60° between full extension and full flexion. Additionally, the gait phase detection method remained robust during this experiment. These results highlight the fact that

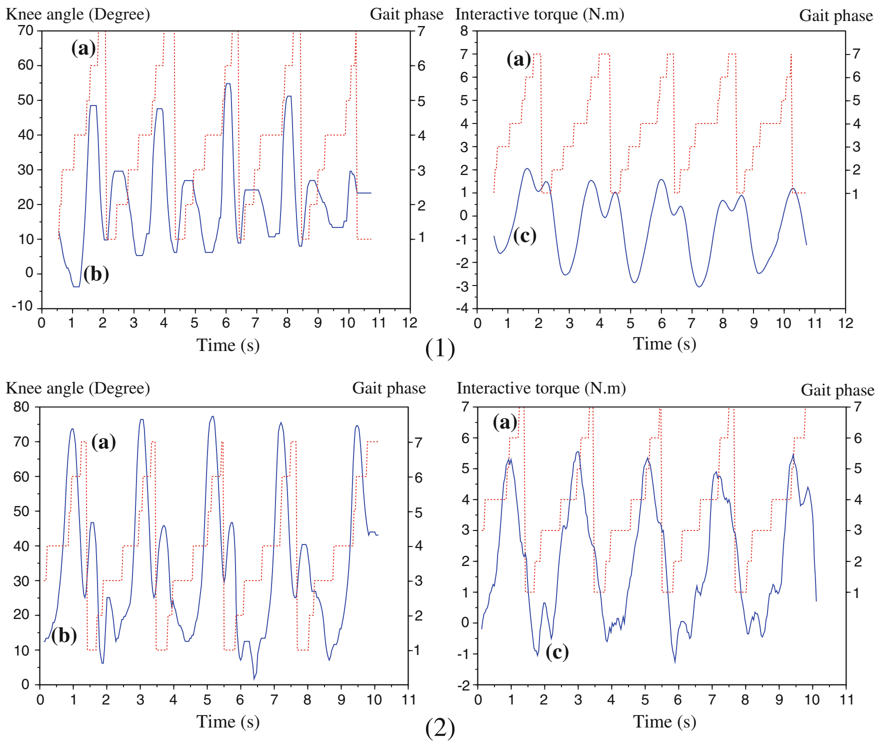


Fig. 4 **1** First experiment: without motor torque control. **2** Second experiment: with predictive control. Curve (*a*): Gait phases. Curve (*b*): Knee angular variation. Curve (*c*): Interactive torque

this method can actually be used to assist the user's knee during a specific gait phase, while allowing the system to remain transparent during the other gait phases.

5 Conclusion

In this paper, the authors describe the control solution for a lower limb exoskeleton, which is designed to provide torque assistance for the knee joints during locomotion. The mechanical design of the prototype, comprising the kinematic solution for the external structure and passive linkages serving as connections between the latter and the user's body, was presented. Solutions for the control of the interaction between the user and the device, as well as for the gait phases detection were also discussed and experimented. The gait phases detection method was realized by using data collected by different displacement sensors placed at the hip joints and the knee joints levels, allowing an estimation in real time. The experiment results highlight the fact that a robust prediction model of the user's gait phases can help improving the control of the system's transparency, as well as the force/torque assistance capacity for a targeted anatomical joint during walking.

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