Control Engineering Practice 26 (2014) 245–253

Contents lists available at ScienceDirect

Control Engineering Practice

journal homepage: www.elsevier.com/locate/conengprac

# Powered orthosis for lower limb movements assistance and rehabilitation

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ARTICLE INFO

*Article history:*
Received 17 August 2013
Accepted 1 February 2014

*Keywords:*
Actuated orthosis
Wearable robots
Musculoskeletal modeling
Rehabilitation robotics

ABSTRACT

This paper presents two rehabilitation approaches of the lower limb: passive and active. The passive one ensures repetitive motions of the limb without any effort delivered by the wearer. Within the active one, a human–exoskeleton interaction approach is proposed. It allows us to provide a knee joint torque support, adapted according to the intention and ability of the wearer, for assistance-as-needed. The wearer's intention is estimated using a realistic model of the muscles actuating the knee joint. The identification process concerns the inertial parameters of the shank-foot-exoskeleton and the musculotendinous parameters. Experiments were conducted online on a healthy subject and have shown satisfactory results in terms of tracking error, intention detection and passive-rehabilitation/active-assistance.



## 1. Introduction

Assistance/rehabilitation robot applications based on human–robot interactions have gained increasing interest over the past decade with the continuous rise of the aged population (Kazerooni, Racine, Huang, & Steger, 2005; Mohammed, Amirat, & Rifai, 2012). Nowadays, the rise in life expectancy is set to continue, combined with falling birth rates, will certainly accelerate the ageing of the population. The EU population aged 60 and over is expected to rise by 37% by 2050.[1] Dependent people (patients and elderly) has limited capabilities to achieve various tasks of the daily life activities. This issue can be surmounted by applying a program of rehabilitation exercises. The use of exoskeletons can offer a solution for the rehabilitation tasks that are often repetitive and require some endurance from both therapist and patient. Exoskeletons are wearable mechanical external devices with actuated joints associated to those of the human body (Pons, 2008). One can notice two major challenges when using exoskeletons: the first one is related to the mechanical aspect and consists in the design of an appropriate mechanical structure that offers good ergonomics and optimal energy transfer to the wearer. The second challenge regards the interaction between the exoskeleton and the wearer. Physical and cognitive interactions between the wearer and the exoskeleton are explored to have an optimal use of the exoskeletons (Pons, 2008). Cognitive interaction can be used to adapt the assisting torque generated by the exoskeleton upon the wearer intention initiated by the Central Nervous System (CNS).

This paper lies in the continuity of previous works (Rifai, Mohammed, Hassani, & Amirat, 2013; Hassani, Mohammed, Rifai, & Amirat, 2013, 2013) done in the same context of controlling a wearable knee joint exoskeleton. In this paper, two approaches of rehabilitation/assistance are studied. The first one deals with the passive rehabilitation where no effort is delivered by the wearer. It consists of performing repetitive motions of the limbs following a desired trajectory predetermined by a rehabilitation doctor. This approach concerns mainly patients having undergone total knee arthroplasty or spinal cord injuries. Within the second approach, a human–exoskeleton interaction strategy is proposed to provide torque assistance upon the wearer intention in order to produce voluntary lower limb movements. The intention of the wearer is estimated by using a realistic musculoskeletal model of the muscles actuating the knee joint as well as measurements of electromyographic (EMG) signals. This musculoskeletal model is based on the use of a modified Hill type muscle model that takes into account the representation of bones and insertion points of the associated muscles. The exoskeleton's torque is adapted according to the intention and abilities of the wearer and can be either resistive or assisting. The rest of the paper is organized as follows: Section 2 discusses the related research works. Section 3 presents the modeling and parametric identification. Section 4 presents the proposed control law used to generate an appropriate exoskeleton torque to track a predefined trajectory or a trajectory

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[1] http://bookshop.europa.eu

http://dx.doi.org/10.1016/j.conengprac.2014.02.002
0967-0661 

determined with respect to the wearer muscular activities. Section 5 shows the experiments conducted on a healthy subject and Section 6 presents the conclusion and the perspectives of this study.

## 2. Related works

Several previous works have dealt with the control of wearable robots for upper and/or lower limbs. A non-exhaustive of related research works is sketched in the following: in Sawicki, Gordon, and Ferris (2005), the authors proposed a one Degree Of Freedom (DOF) wearable powered ankle foot orthosis with artificial pneumatic muscle for gait rehabilitation. The control scheme is based on the assisting torque estimation by using a proportional controller with respect to the EMG activation level. An EMG interface to control a two degrees of freedom exoskeleton is presented in Kiguchi, Iwami, Yasuda, Watanabe, and Fukuda (2003). This prototype provides assistance for flexion/extension and pronation/supination of the shoulder joint while performing predefined movements. The controller output is only moderated through raw EMG patterns previously learned using a fuzzy-neuro model. Authors in Khokhar, Xiao, and Menon (2010) used a raw EMG pattern recognition approach to control a two DOF wrist exoskeleton. A support vector machine is used to classify the EMG signals and to estimate the wearer's intention. The estimated torque is shared linearly to assist the wearer's movements. The Hybrid Assistive Limb (HAL) relies on the detection of motion intention and the achievement of the movement task. The so-called voluntary control system estimates the wearer's intention through the detection of the EMG signals (Kawabata, Satoh, & Sankai, 2009; Masahiro, Kiyoshi, & Yoshiyuki, 2009). In Rosen, Brand, Fuchs, and Arcan (2001) and Rosen, Fuchs, and Arcan (1999), a fixed upper limb exoskeleton is used to assist flexion and extension movements of the elbow joint. The wearer's force is evaluated using EMG with a simplified Hill-type muscle model. The proposed strategy is based on the use of neural networks. The assistive torque provided by the exoskeleton is linearly dependent on the wearer's torque. However, the parametric identification as well as the experimental validation of the musculoskeletal model were not provided. In Fleischer and Hommel (2008), a knee joint exoskeleton controlled through the wearer's intention estimation is proposed. A musculoskeletal knee joint model is introduced to improve the estimation of the wearer's torque. Two control schemes were used. In the first one, the estimated wearer's torque is linearly dependent on the assisting torque, while in the second one, the wearer's movement is estimated using the knee joint torque of the wearer and its inverse rigid-body dynamics.

## 3. Modeling and identification

### 3.1. Wearer-exoskeleton modeling

The system is composed of the human shank–foot and the embodied actuated exoskeleton that fits perfectly the wearer's leg to which it is fixed by means of straps. The exoskeleton is driven by an actuator and the whole system is in a synchronized motion. The exoskeleton is also subjected to the human effort delivered by the muscles of the lower limb acting on the knee joint level. The system is intended to assist flexion–extension movements of the human shank–foot for rehabilitation purposes.

The wearer is supposed to be in a sitting position performing flexion and extension movements of the knee joint without taking into account the ground contact (Fig. 1). The model of the system composed of the exoskeleton and the human shank–foot is given

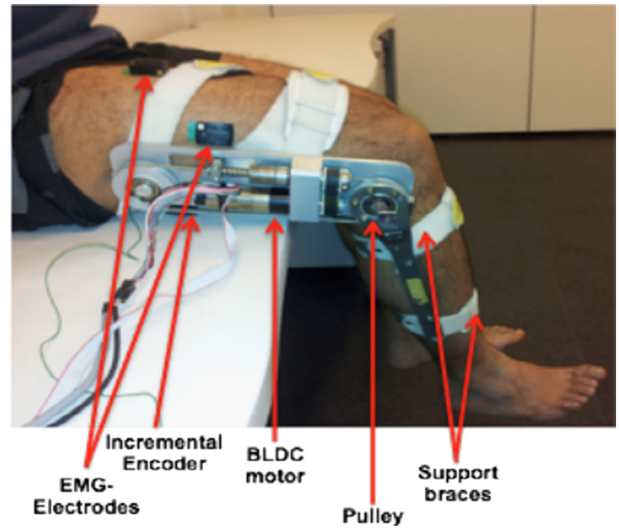


**Fig. 1.** A person wearing the exoskeleton in a sitting position.

by the following:

$$J\ddot{\theta}+B\dot{\theta}+\frac{\partial E_p}{\partial \theta}=\tau_e+\tau_h \tag{1}$$

where $\theta$, $\dot{\theta}$, and $\ddot{\theta}$ are the knee joint angle, angular velocity and acceleration respectively. $J$ is the system's moment of inertia, $B$ is the system's damping, $E_p$ is the potential energy given by the following:

$$E_p=\frac{1}{2}K(\theta-\theta_r)^2-\tau_g\ \cos(\theta-\theta_r) \tag{2}$$

with $K$ being the knee joint stiffness, $\tau_g$ being the system's gravitational torque at the full extension position of the shank, $\theta_r$ being the knee joint angle at the rest position. $\tau_e$ is the torque delivered by the exoskeleton's actuator and $\tau_h$ is the torque delivered by the lower limb muscles and is given by the following (Fig. 3):

$$\tau_h=\sum_{j=1}^{N_m}F_j^{mt}(EMG_j,\theta)ma_j \tag{3}$$

where $F_j^{mt}$ is the force of the $j$th musculotendon and $ma_j$ is its moment arm, $N_m$ is the number of musculotendons.

The musculotendon model is based on a modified Hill-type muscle model. A musculotendon unit consists of a contractile element in parallel with a damping component and in series with a nonlinear elastic tendon (Schutte, 1993). The musculotendon unit force $F^{mt}$ is equal to the tendon force $F^t$ and is given as follows:

$$F^{mt}(t)=F^t(t)=F_{max}[a(t)f_l(\bar{l}^m(t))f_v(\bar{v}^m(t))+\bar{b}^m\bar{v}^m(t)] \tag{4}$$

where

$$\dot{a}(t)+\frac{1}{\tau_{act}}\left[\frac{\tau_{act}}{\tau_{deact}}+\left(1-\frac{\tau_{act}}{\tau_{deact}}\right)c(t)\right]a(t)=\frac{1}{\tau_{act}}c(t) \tag{5}$$

$$f_l(\bar{l}_m(t))=e^{-(\bar{l}^m(t)-1)^2}/\gamma$$

$$f_v(\bar{v}^m(t))=\begin{cases}\dfrac{k_v[\bar{v}^m(t)+1]}{-\bar{v}^m(t)+k_v}, & \bar{v}^m(t)\le 0\\[2ex] \dfrac{1.8[k_v+1]\bar{v}^m(t)+0.13k_v}{[k_v+1]\bar{v}^m(t)+0.13k_v}, & \bar{v}^m(t)>0\end{cases} \tag{6}$$