Control Engineering Practice 80 (2018) 49–60

Contents lists available at ScienceDirect

Control Engineering Practice

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

Robust Active Disturbance Rejection Control via Control Lyapunov Functions: Application to Actuated-Ankle–Foot-Orthosis

J.F. Guerrero-Castellanos[a,*], H. Rifaï[b], V. Arnez-Paniagua[b], J. Linares-Flores[c,1], L. Saynes-Torres[a], S. Mohammed[b]

[a] *Facultad de Ciencias de la Electrónica, Benemérita Universidad Autónoma de Puebla (BUAP), Ciudad Universitaria, C. P. 72570, Puebla, Mexico*
[b] *Laboratoire Images, Signaux et Systèmes Intelligents (LISSI), Université Paris-Est Créteil (UPEC), 94400 Vitry-sur-Seine, France*
[c] *Universidad Tecnológica de la Mixteca, Oaxaca 69000, Mexico*

ARTICLE INFO

Keywords:
Control Lyapunov Function (CLF)
Actuated-Ankle–Foot-Orthosis (AAFO)
Active Disturbance Rejection Control (ADRC)
Extended State Observer (ESO)
Input-to-State Stability (ISS)

ABSTRACT

In this paper, the problem of trajectory tracking, for an Actuated-Ankle–Foot-Orthosis (AAFO) to assist the gait of paretic patients, is addressed. The control strategy is based on the system's flatness property, which allows the development of an Active Disturbance Rejection Control (ADRC). For this purpose, an Extended State Observer (ESO) is designed to estimate on-line the unknown disturbances and canceled by injecting the output of ESO into the feedback loop. A stability analysis of the estimation error dynamics is carried out in the Input-to State Stability (ISS) framework, stating the observer's robustness. On the other hand, the feedback design is based on the existence of a Control Lyapunov Function (CLF) and the Sontag's formula. The stability analysis discloses that the tracking trajectory controller is ISS, *i.e.* robustly stable. Finally, the effectiveness of the ADRC strategy is validated by performing real-time tests with a healthy subject walking on a treadmill at self-selected speed. The experimental results validate the theoretical analysis.

1. Introduction

1.1. Motivations and background

Neurological injuries such as Spinal Cord Injuries (SCI) and stroke, results frequently in Foot-Drop symptom (Stewart, 2008). Indeed, every year an augmenting number of people are diagnosed with disabilities that prevent them from performing daily living activities such as walking, stairs ascent/descent, standing up, etc. According to the World Health Organization, around 15% of the world's population lives with some form of disability; among them 2%–4% are facing important difficulties while performing daily activities. Patients suffering from gait pathologies may either have total loss of ankle muscle forces and are unable to initiate a movement with the affected limbs or they may have partial loss of muscle forces and are able to move their limbs within limited ranges. Furthermore, spasticity can occur during a gait cycle due to the involuntary contraction reflex of antagonist muscles spanning at the ankle joint. The most common gait pathologies at the ankle joint is foot-drop, foot slap, and insufficient push-off power. Foot-drop patients are unable to lift their feet and toes properly during walking, affecting thus their movement coordination, their balance and increasing the risk of falling. Treatments for this pathology range from conventional therapy, use of passive orthoses, functional electrical stimulation (FES) (Coste, Jovic, Pissard-Gibollet, & Froger, 2014; Peckham & Knutson, 2005), to the use of wearable robotics based solutions, known also as actuated ankle–foot orthosis (AAFO).

Conventional therapy include lower limb muscle strengthening exercises, joint stretching to enhance ankle dorsiflexion and plantarflexion, and ground walking with the assistance of therapists. It is worth noting, however, that this therapy is difficult and effort demanding to be performed continuously for more than few minutes by both therapists and patients (Jamwal, 2011; Krishnamoorthy, Hsu, Kesar, Benoit, Banala, Perumal, Sangwan, Binder-Macleod, Agrawal, & Scholz, 2008; Nicholas Romansky, Kelly Scollon-Grieve, & McGinness, 2012). Therefore, the inclusion of robotic devices such as AAFOs could potentially increase the dosage and intensity of the therapy while reducing the effort required from the therapists. In robotic-assisted therapy, some examples of AAFOs used to prevent foot-drop provide assistance torque in the dorsiflexion direction (Roy, Krebs, Iqbal, Macko, Macko,

* Corresponding author.
E-mail address: fermi.guerrero@correo.buap.mx (J.F. Guerrero-Castellanos).
URL: http://www.fguerrero.ece.buap.mx (J.F. Guerrero-Castellanos).
[1] On sabbatical leave at Benemérita Universidad Autónoma de Puebla (BUAP).

https://doi.org/10.1016/j.conengprac.2018.08.008
Received 2 March 2018; Received in revised form 9 June 2018; Accepted 14 August 2018
0967-0661/

& Forrester, 2014). However, another advantage of actuated orthosis compared to passive ones is the capacity of the former group to provide assistance in the plantar-flexion direction to promote a better foot push-off power prior to the swing phase (Blaya & Herr, 2004; Park, Chen, Young, Stirling, Wood, Goldfield, & Nagpal, 2011). There are several strategies to determine the assistance level provided by the AAFOs (Boehler, Hollander, Sugar, & Shin, 2008; Jimenez-Fabian & Verlinden, 2011; Shorter, Xia, Hsiao-Wecksler, Durfee, & Kogler, 2013): (1) by pre-selecting the assistance torque and applying it in a feedforward scheme with respect to the gait phase detected (Ab Patar, Said, Mahmud, Majeed, & Razman, 2014; Arnez-Paniagua, Huo, Colorado-Cervantes, Mohammed, & Amirat, 2016; Boehler et al., 2008; Jamwal, 2011; Shorter, Kogler, Loth, Durfee, & Hsiao-Wecksler, 2011), (2) as a function of electromyography (EMG) signals (Ferris, Gordon, Sawicki, & Peethambaran, 2006; Pérez-Ibarra & Siqueira, 2017), (3) by adapting the stiffness, inertia or impedance of the coupled human-exoskeleton system based on the gait phase detected, (Blaya & Herr, 2004; Lawn, Takashima, Ninomiya, Yu, Soma, & Ishimatsu, 2015; Roy et al., 2014; el Zahraa Wehbi, Huo, Amirat, El Rafei, Khalil, & Mohammed, 2017), or (4) as a function of the tracking error between the current ankle joint angle and a reference trajectory pattern commonly generated from healthy subject walking profiles (Arnez-Paniagua, Rifaï, Amirat, & Mohammed, 2017a; Bharadwaj, Sugar, Koeneman, & Koeneman, 2005; Brahmi, Saad, Ochoa-Luna, & Rahman, 2017; Hitt, Oymagil, Sugar, Hollander, Boehler, & Fleeger, 2007; Holgate, Bohler, & Suga, 2008; Jamwal, Xie, Hussain, & Parsons, 2014; Madani, Daachi, & Djouani, 2014; Rifaï, Mohammed, Hassani, & Amirat, 2013; Veneva & Ferreira, 2014; Ward, Sugar, Standeven, & Engsberg, 2010; Wolbrecht, Chan, Le, Cramer, Reinkensmeyer, & Bobrow, 2007; Zhang, Cao, Xie, Zhu, Zeng, Huang, & Xu, 2017). The control method used for each exoskeleton has a direct impact on the level and rate of human adaptation to the active device; i.e., the nervous system can adapt more easily to a continuous, smooth and proportionate stimulus (Cain, Gordon, & Ferris, 2007). In this sense, it is clear that there exist an intimal relation between AAFO and the control strategies design based on rehabilitation objectives (Jiménez-Fabián & Verlinden, 2012). Feedforward strategies are simple to implement but the lack of feedback could produce an insufficient or excessive assistance. EMG based strategies provide a high rate of adaptation for the nervous system, but requires a minimum residual muscular activities in order to be effective, which might not be the case for some patients with acute stroke symptoms. The systems that adapt their impedance have the advantage of not requiring a predefined trajectory but they require a minimum residual voluntary effort to initiate movements. Therefore, a trajectory tracking strategy could potentially provide more appropriate assistance when the wearer of the active orthosis is not fully able to initiate movement.

The present paper is more in the spirit of the fourth group. Since AAFO is modeled as a nonlinear system subject to disturbances from the external environment and uncertainties such as unmodeled dynamics, parameter perturbations, and nonlinear couplings, the control problem for gait tracking to achieve ankle rehabilitation is in itself a challenge (Gregg, Bretl, & Spong, 2010). Proportional control (also called High-Gain Control) and PID control results in an attractive solution due to its simplicity to real-time implementation (Åström & Murray, 2008) and it has been applied to AAFO, for instance in Boiadjiev and Veneva (2009) and Bai, Gao, Zhao, Jin, Dai, and Lv (2015). However, it is well known that proportional control cannot completely remove the effects caused by disturbance (even a constant one) and a higher control gain has to be designed to diminish the disturbance effects. On the one hand, when PID is used, the integral term could effectively reject the constant disturbance but achieves poor performance in the presence of time-varying disturbances. On the other hand, PID is often implemented without the D part because of the noise sensitivity (Han, 2009). Adaptive Control (AC) is quite effective in dealing with model parameter uncertainties and has gained wide applications in practical engineering. The successful applications of AC methods usually depend highly on the design of identification or estimation laws on time-varying model parameters (Krstic, Kanellakopoulos, & Kokotovic, 1995) and has been recently applied to AAFO in Ibarra, dos Santos, Krebs, and Siqueira (2014) and Arnez-Paniagua, Rifai, Mohammed, and Amirat (2017b). However, when external disturbances are present, e.g. muscular torque due to gait pathologies or the impact with the ground, AC is limited since the error produced by the disturbance could negatively affect the adaptive performance and key parameters are difficult to identify or estimate online (Arnez-Paniagua et al., 2017b). Sliding Model Control (SMC) has fine abilities in suppressing the effects of parameter perturbations as well as external disturbances (Spurgeon, 2014). However, it is well known that the discontinuous switching of the controller is prone to induce high-frequency chattering of mechanical systems which would be unacceptable for the AAFO. The employment of some modification methods could effectively reduce the chattering problem (Bartolini, Pisano, Punta, & Usai, 2003), as has been shown recently in Mohammed, Huo, Huang, Rifa, and Amirat (2016) for a knee joint orthosis. In robotics field, impedance control (IC) has been largely used for manipulator robots (Hogan, 1985). IC establishes a relationship between the force, the velocity and the environment. In the last years, remarkable works have been proposed to deal with simple, safe and robust motion/force tracking controllers (Mehdi & Boubaker, 2012a, b) and applied to rehabilitation robotic systems with parameters uncertainties (Mehdi & Boubaker, 2016). Furthermore, when the contact with the ground is taken into account, Jerk-Impedance (Aloulou & Boubaker, 2015) and Jerk-Stiffness (Aloulou & Boubaker, 2016) controllers have been designed for gait pattern generation and safe walking with applicability to a breadth of rehabilitation robotics applications.

In the context of nonlinear control systems, since the seminal works of Artstein and Sontag (Sontag, 1998), Control Lyapunov Functions, so called CLFs, have become central to feedback design. A main reason is that the existence of a CLF is necessary and sufficient for the stabilizability of a system with a control input. Domains of application include robust nonlinear feedback design (Sepulchre, Jankovic, & Kokotović, 1997), receding horizon control of nonlinear systems (Primbs, Nevistic, & Doyle, 1999), stabilization of hybrid systems (Sanfelice, 2013) and stabilization of nonlinear system with event-based control (Marchand, Durand, & Guerrero-Castellanos, 2013), to name only a few. In the bipedal walking robot framework, CLF approach has been successful used to exponentially stabilize periodic orbits of the hybrid zero dynamics by shaping the energy (Ames, Galloway, Sreenath, & Grizzle, 2014; Galloway, Sreenath, Ames, & Grizzle, 2015), where the control laws are based on the Sontag's formula which is well known to possess robustness to static and dynamic input uncertainties (Jankovic, Sepulchre, & Kokotović, 1999). Former properties represent a main motivation to use CLFs in the present work.

The objective of the above-mentioned control approaches is to reject disturbances via feedback, which is based on the tracking error between the measured outputs and their setpoints or desired trajectories. As a consequence, these controllers cannot react fast enough in the presence of strong disturbances. In order to overcome this limitation, Active Disturbance Rejection Control (ADRC) was introduced by Han (2009). ADRC is fundamentally based on the possibility of on-line estimating adverse effects so called "total disturbance" caused by the coupling between unknown system dynamics (endogenous) and external (exogenous) disturbances. This estimation is then canceled via an appropriate feedback-feedforward control law (Sira-Ramírez, Luviano-Juárez, Ramírez-Neria, & Zurita-Bustamante, 2017). The most remarkable feature of ADRC lies in its estimation/cancellation nature, where the total disturbance is considered as an extended state and is estimated, in real-time, through an Extended State Observer (ESO) (Chan, Naghdy, & Stirling, 2013; Mehdi & Boubaker, 2015; Xue, Huang, & Gao, 2016) so called disturbance-observer (Yu, Yang, Han, & Liu, 2018). ADRC has been exploited in almost all domains of control engineering for example: motion control of humanoid robots (Orozco-Soto & Ibarra-Zannatha, 2017), power filter design (Fuentes, Cortés-Romero, Zou, Costa-Castelló, & Zhou, 2015), energy storage (Chang, Li, Zhang, Wang, &