Augmented L1 adaptive control of an actuated knee joint exoskeleton: From design to real-time experiments
Hala Rifaï, Mohamed Ben Abdessalem, Ahmed Chemori, Samer Mohammed, Yacine Amirat

To cite this version:

HAL Id: lirmm-01723920
https://hal-lirmm.cnrs.fr/lirmm-01723920
Submitted on 5 Mar 2018

HAL is a multi-disciplinary open access archive for the deposit and dissemination of scientific research documents, whether they are published or not. The documents may come from teaching and research institutions in France or abroad, or from public or private research centers.

L’archive ouverte pluridisciplinaire HAL, est destinée au dépôt et à la diffusion de documents scientifiques de niveau recherche, publiés ou non, émanant des établissements d’enseignement et de recherche français ou étrangers, des laboratoires publics ou privés.
Augmented $L^1$ Adaptive Control of an Actuated Knee Joint Exoskeleton: From Design to Real-Time Experiments

H. Rifaï, M.S. Ben Abdessalem, A. Chemori, S. Mohammed and Y. Amirat

Abstract—This paper deals with the control of a lower limb exoskeleton acting on the knee joint level. Classical $L^1$ adaptive control law is proposed to ensure assistance-as-needed and resistive rehabilitation following a desired trajectory considered defined by a therapeutic doctor. This control law introduces a time lag within the desired trajectory tracking because of the presence of a filter in its structure. In order to mitigate this drawback, the classical $L^1$ adaptive control is augmented by a nonlinear proportional control. The classical and augmented $L^1$ adaptive control laws are tested in real-time using the exoskeleton EICOSI of LISSI-lab. Real-time experimental results highlight the utility of these control laws in assistance-as-needed and resistive rehabilitation. They also show the effectiveness of the augmented version of the $L^1$ adaptive control.

I. INTRODUCTION

The ageing of the societies and the high percent of disabilities have encouraged the development of exoskeletons [1]. These external skeletal systems can be used in rehabilitation but also in performance enhancement. Within the first application, they enable the wearer to regain control of his/her limbs and to perform movements that have been lost [2]. Within the second application, the exoskeletons empower the wearer and allow him to perform tasks of high endurance such as load carrying [3].

Lower limb exoskeletons act particularly on the hip, knee and ankle joints or separately on these articulations. They aim to assist the movement of the lower limbs at the impairment level during daily living activities. When the wearer develops a part of the effort necessary to perform a movement and the exoskeleton delivers the complement, the paradigm is called assistance-as-needed. The lower limb exoskeletons aim also to perform rehabilitation programs trying to restore the movement of the impaired joints. Within this context, two programs can be envisaged: passive and resistive rehabilitations. Passive rehabilitation concerns people who have completely lost control of their limbs after spinal cord injuries for example [4], [5]. The resistive rehabilitation, on the other hand, involves the wearer in the process by inviting him to develop an effort in the opposite direction of the exoskeleton’s movement [6]. In both cases, the rehabilitation process is monitored by a therapist whose workload is drastically alleviated thanks to the automated process.

Different control techniques have been established to automatically drive the lower limb exoskeletons. One can cite the classical PID (proportional, integral, derivative) structure for gait assistance and its derivatives such as the proportional controller which acts as an amplifier of the human developed torque estimated using EMG (electromyogram) electrodes measurements [7], [8] and the proportional derivative controller where the wearer force is measured through shoes insoles [9]. In [2], a position and torque controllers that generate a torque proportional to the deviation angle from the desired trajectory are proposed for gait rehabilitation of hemiparetic patients, the control torque is applied only when the wearer needs to be assisted. A bounded control law that has the advantage of avoiding the saturation of the exoskeleton’s actuator has been proposed in [10] to ensure passive rehabilitation of a completely disabled wearer, it allows to guarantee the security of the wearer by maintaining the functioning of the actuator in a linear mode. A quasi-passive structure of exoskeleton is presented in [11] where the stiffness of the knee is on-off controlled with respect to stance phases detected with shoe insoles. Impedance control has been also developed for exoskeletons such that the impedance of the exoskeleton-wearer system can be adapted acting on the inertia, damping or stiffness parameters [12], [13], [14]. Adaptive control has been proposed in [15] allowing to adapt all the system’s (exoskeleton and wearer) parameters function of the task to perform. Adaptive strategies allow to ensure position control with a compliant wearer-exoskeleton system: the parameters are adapted continuously so that the best performance in desired trajectory tracking is ensured. Note that compliant structures of exoskeletons have been developed such that the exoskeleton’s impedance can vary from very stiff to very compliant depending on the task to perform [16].

Thanks to the aforementioned advantages of the adaptive control, a $L^1$ adaptive position controller will be proposed in this work. Besides the parameters adaptability, the $L^1$ adaptive control ensures that the signals in the feedback loop remain bounded due to the projection function existing in its expression [17]. Moreover, it guarantees a transient performance and robustness in presence of parameters’ fast adaptation. Robustness and adaptation are decoupled and a compromise can be set by means of a filter present in the control law structure. This filter, however, is behind the main drawback of the $L^1$ adaptive control because it creates a time lag within the tracking of the desired trajectory.
This drawback can be mitigated by augmenting the $L_1$ adaptive control with a nonlinear proportional or a PI, PID controllers as proposed in [18], [19]. Among applications of the $L_1$ adaptive control, one can cite the aerial vehicles, the underwater vehicles, the robot manipulators, etc. [19], [20], [21].

To the authors best knowledge, the $L_1$ adaptive control has not been applied to wearable robots previously. In this paper, the $L_1$ adaptive control is applied to the Exoskeleton Intelligently Communicating and Sensitive to Intention (EICOSI) that has a rotational degree of freedom at the knee joint level [10].

The remainder of the paper is structured as follows. In section II, the shank-foot-exoskeleton model is recalled. The classical and augmented $L_1$ adaptive control laws are presented in section III. Real-time experiments using EICOSI are addressed in section IV for assistance-as-needed and resistive rehabilitation. Finally, conclusions and future works are sketched in section V.

II. SHANK-FOOT-EXOSKELETON: DESCRIPTION AND MODELING

EICOSI (cf. Fig. 1) is a lower limb exoskeleton that has a rotational degree of freedom at the knee joint level. It is driven by a brushless DC motor and is equipped with an incremental encoder to measure the knee joint angle. It is embodied by the lower limb and attached to by means of straps such that it rotates synchronously with the knee joint. The shank-foot-exoskeleton is therefore considered as an entity that is put in movement thanks to the exoskeleton actuator’s torque and to the wearer’s active muscular torque if present. These two torques are the shank-foot-exoskeleton system’s external inputs. The wearer is considered in a seated position, the shank is hanging (cf. Fig. 4). Flexion and extension movements of the knee joint are studied in this work such that the system tracks a desired trajectory considered defined by a therapeutic doctor. Deriving the Lagrangian of the shank-foot-exoskeleton system, the rotational dynamics can be obtained [10]:

$$J\ddot{\theta} = -\tau_g \cos \theta - K(\theta - \theta_r) - A \text{sign} \dot{\theta} - B \dot{\theta} + \tau + \tau_h,$$

where $\theta$ is the knee joint rotation angle which is defined as the angle between the actual position of the shank and the full extension position, $\dot{\theta}$ and $\ddot{\theta}$ are the knee joint angular velocity and acceleration respectively, $\theta_r$ is the angle of the knee joint at the rest position of the shank, $J$ is the shank-foot-exoskeleton system’s moment of inertia, $\tau_g$ is the system’s gravitational torque in the full extension position of the shank, $K$ is the system’s stiffness coefficient, $A$ is the system’s solid friction coefficient, $B$ is the system’s viscous friction coefficient and $\text{sign}(\cdot)$ is a signum function. $\tau$ is the control torque delivered by the exoskeleton’s actuator and $\tau_h$ is the human active torque generated by his active muscles.

III. PROPOSED CONTROL SOLUTION: AUGMENTED $L_1$ ADAPTIVE CONTROL

A. Background on $L_1$ adaptive control

In order to avoid degradation in the closed-loop performance of a controlled system, adaptive control has been developed. Indeed, the controller is expected to possess a self tuning ability and compensate for different kind of uncertainties/disturbances and changes in the system/environment. That is why adaptive controllers are very popular for such systems. However, it is worth to emphasize various issues related to the implementation of a classical adaptive controller, such as i) the need for the persistency in excitation that can lead to a bad transient behavior, ii) the tuning of the adaptation gains that can lead either to instability (high gains) or slow down the convergence rate (small gains), iii) the need of an appropriate initialization of the estimated parameters requiring an a priori knowledge of the system. The $L_1$ adaptive control [17] scheme stands out among all other developed methods in its particular architecture where robustness and adaptation are decoupled. Its architecture is constructed (inspired by Model Reference Adaptive Control) using four main parts as illustrated in Fig. 2, namely the controlled system, the state predictor, the adaptation mechanism and the control law including a low pass filter.

![Fig. 1. View of the exoskeleton system and its components.](image1)

![Fig. 2. View of the block diagram of $L_1$ adaptive control scheme](image2)
The controlled system: Let’s consider the class of systems modelled by the following dynamics:

\[
\dot{x}(t) = Ax(t) + b\left(\omega(t)u(t) + W^T(t)x(t) + \sigma(t)\right), \quad x(0) = x_0
\]

\[
y(t) = cx(t)
\]

(2)

\(x \in \mathbb{R}^n\) is the state vector which is supposed to be measurable, \(u(t) \in \mathbb{R}\) is the control input, \(b \in \mathbb{R}^n\) and \(c \in \mathbb{R}^{1 \times n}\) are constant vectors supposed to be unknown, \(A \in \mathbb{R}^{n \times n}\) is a known matrix such that \((A, b)\) is controllable, \(W \in \mathbb{R}^{n^2}\) is the vector of unknown constant parameters with a known bound. \(\sigma(t) \in \mathbb{R}\) is a time varying disturbance, while \(\omega(t) \in \mathbb{R}\) is unknown constant with known sign, \(y\) is the system’s output. Having defined a piecewise-continuous bounded reference trajectory \(r(t)\), the control objective lies in the design of an adaptive state feedback input \(u(t)\) so that the output \(y(t)\) tracks \(r(t)\) while preserving the boundedness of the states and the parameters of the system. This control input is composed of two parts defined according to the following:

\[
u(t) = u_m(t) + u_a(t) \quad , \quad u_m(t) = -k_m x(t)
\]

(3)

\(u_m\) is the component rendering the matrix \(A\) Hurwitz thanks to \(k_m \in \mathbb{R}^{1 \times n}\) being the static gain factor that transforms \(A\) into \(A_m = A - bk_m\). The matrix \(A_m\) is therefore the one delimiting the closed-loop dynamics of the system. It remains \(u_a\) being the adaptive control input to be designed. By combining (2) and (3), one gets:

\[
\dot{x}(t) = A_m x(t) + b\left(\omega(t)u_a(t) + W^T(t)x(t) + \sigma(t)\right), \quad x(0) = x_0
\]

\[
y(t) = cx(t)
\]

(4)

The matrix \(A\) is replaced by the Hurwitz one \(A_m\) thanks to the incorporation of \(u_m(t)\). The control input is therefore reduced to the adaptive component \(u_a(t)\).

The state predictor: The states of the system are computed at every iteration based on the estimated parameters obtained from the adaptation mechanism along with the control input. The predictor is therefore constructed according to the following:

\[
\dot{x}(t) = A_m \hat{x}(t) + b\left(\hat{\omega}(t)u_a(t) + \hat{W}^T(t)x(t) + \hat{\sigma}(t)\right), \quad x(0) = x_0
\]

\[
\hat{y}(t) = c\hat{x}(t)
\]

(5)

\(\hat{x}(t) \in \mathbb{R}^n\) is the state of the predictor of the system and \(\hat{W}^T(t) \in \mathbb{R}^n\) is the estimate of the vector of unknown constant parameters. Finally \(\hat{y} \in \mathbb{R}\) is the estimate of the output.

The adaptation mechanism: It uses the error between the measured and the estimated states to adapt the parameters together with a projection method in order to ensure their boundedness. Indeed, a projection operator avoids the parameter drift using the gradient of a convex function and a maximal bound on the parameters to be estimated. The adaptation law for each estimated parameter is then given by:

\[
\dot{\hat{W}}(t) = \Gamma W \text{Proj}(\hat{W}(t), -\hat{\omega}(t)^T(t)Ph(t))
\]

\[
\hat{\sigma}(t) = \Gamma_\sigma \text{Proj}(\hat{\sigma}(t), -\hat{\omega}(t)^T(t)Ph)
\]

\[
\dot{\hat{\omega}}(t) = \Gamma_\omega \text{Proj}(\hat{\omega}(t), -\hat{\omega}(t)^T(t)Ph u_a(t))
\]

(6)

where \(\hat{W}\) is the solution of the algebraic Lyapunov equation \(A_m^T P + P A_m^T = -Q\) for any arbitrary symmetric matrix \(Q = Q^T > 0\). \(\Gamma_W, \Gamma_\sigma, \Gamma_\omega\) are the adaptation gains and \(\hat{\omega}(t)\) is the prediction error.

The control law with low pass filter: This part pertains to the adaptive component of the control input characterized by the consideration of a low pass filter. It is written in Laplace form as:

\[
u_a(s) = -C(s)(\hat{\eta}(s) - k_g r(s))
\]

(7)

where \(C(s)\) is a bounded input, bounded output stable and strictly proper transfer function, \(\hat{\eta}(s) = \hat{W}(s)x(s)\), \(k_g = \frac{1}{\hat{c}A_m b}\) and \(r(s)\) the Laplace transform of the reference trajectory \(r(t)\). The system (2) under the \(\mathcal{L}_1\) adaptive controller proposed by (7) is guaranteed to be bounded input bounded state stable with respect to trajectory and initial conditions if \(k_m\) and \(C(s)\) verify with following \(\mathcal{L}_1\) norm condition:

\[
\|G(s)\|_{\mathcal{L}_1} L < 1
\]

(8)

where \(G(s) = H(s)(1 - C(s)), \quad H(s) = (sI - A_m)^{-1}\) with \(L = \max_i|\theta|\)

(9)

\(G(s)\) and \(H(s)\) being bounded-input bounded-output stable transfer functions. \(L\) is the maximal bound set on the parameter \(W\).

Applied to the exoskeleton EICOSI, one can write in case of a passive wearer, then developing a null active muscular torque \((\tau_h = 0)\): \(x = [x_1 \ x_2]^T = [\theta \ \theta]^T, \quad b = [0 \ 1]^T, \quad A_m = \left[\begin{array}{cc} 0 & 1 \\ -k_{m_1} & -k_{m_2} \end{array}\right], \quad W = [-\frac{\omega}{J} + k_{m_1} - \frac{\omega}{J} + k_{m_2}]^T, \quad \omega = \frac{1}{\tau}, \quad \sigma = \frac{J}{\tau} \theta_e - \frac{J}{\tau} \cos x_1 - \frac{4}{3} \sin x_2 \quad c = [1 \ 0].\)

In case of an active wearer \((\tau_h \neq 0), \quad \tau_h\) can be considered as an external disturbance and takes part of the disturbance term \(\sigma\) without changing the proof as it is stated in [22]. The only condition is that \(\tau_h\) should be a continuously differentiable bounded with uniform bounded derivative which is validated since \(\tau_h\) is a continuous muscular bounded torque: \(\tau_h \leq \Delta_h\) where \(\Delta_h\) is the bound.

B. Augmented \(\mathcal{L}_1\) adaptive control

To deal with the issue of time lag, which appears when the controller is applied to track a time-varying reference trajectory, as in [18], an augmentation of the \(\mathcal{L}_1\) adaptive control is proposed. The augmented controller is illustrated in Fig. 3. The extension is displayed in dotted lines. A linear PI controller or a nonlinear proportional controller can be used as the additional term to be summed to the original filtered control input. The idea of the nonlinear proportional control term lies in choosing a nonlinear varying feedback gain. To get a fast transient with a small overshoot, the added control input \(u_p(t)\) can chosen be, according to [23], as follows:

\[
u_p(t) = -g(e, \alpha, \delta)
\]

(10)
with

\[ g(e, \alpha, \delta) = \begin{cases} a_1 |e|^{\alpha} \text{sgn}(e) & \text{if } |e| > \delta \\ a_2 \frac{\alpha}{e^{\alpha}} & \text{if } |e| \leq \delta \end{cases} \]

the parameters \( a_1 \), \( a_2 \) are constant gains (they can be chosen as \( a_1 = a_2 \) for avoiding a discontinuity), \( e \) is the tracking error, \( \alpha \) is a design parameter such that \( 0 < \alpha \leq 1 \) and \( \delta \) is a threshold of the transition between low and high gains.

IV. REAL-TIME EXPERIMENTAL RESULTS

A. Experimental platform and Hardware aspects

The experiments are performed using the Exoskeleton Intelligently Communicating and Sensitive to Intention (EICOSI) (cf. Fig. 1). It is a lower limb exoskeleton, actuated at the knee joint level by means of a brushless DC motor. It is fixed to the wearer’s leg by means of straps that attach it to the thigh and shank. EICOSI and the wearer (cf. Fig. 4) forms then an entity that moves synchronously due to two major active torques: the control torque delivered by the EICOSI’s actuator and the wearer’s torque developed by active thigh muscles. The quadriceps represents the extensor muscles and the hamstrings represents the flexor ones. EMG electrodes are therefore fixed at the level of the rectus femoris (RF) acting as a quadriceps muscle and the biceps femoris long head (BF) acting as hamstring muscle to show when the muscles are active during the experiments. Note that the EMG measurements are presented only to show the muscles activity and are not used in the control law computation. EICOSI is equipped with an incremental encoder that measures the rotational angle of the knee joint. The angular velocity is derived numerically. A dSPACE controller board is used to compute the pulse width modulation level to control the actuator’s velocity taking as inputs the current and desired knee joint angles as well as the angular velocities. Experiments have been performed on a male subject being 23 years old, weighing 65 Kg and measuring 178 cm (cf. Fig. 4). The subject presents no spasticity or contracture of the knee joint and is therefore able to perform complete flexion and extension. He has been informed of the experiment’s protocol and benefit of the study. Precautions have been taken such that the experiments preserve the subject’s health. Privacy and confidentiality have been respected.

All the estimated parameters of the control torque have been initialized to zero. The parameters of the state feedback are taken as: \( k_{m_1} = 1 \) and \( k_{m_2} = 1.4 \).
joint angles. Once the wearer becomes active, the estimated parameters $W_1$ and $W_2$ present a sign alteration (cf. Fig. 6).

C. Resistive rehabilitation

The last two experiments can be divided into two phases: the first one presents a passive wearer who is not delivering any active muscular effort and the second phase presents a wearer who is resisting to the exoskeleton’s movement towards the desired trajectory.
During the third experiment, the classical $\mathcal{L}_1$ adaptive control is implemented. The wearer is passive in the time interval $[0,55]$s and is resisting the exoskeleton's movement towards the desired trajectory after (cf. Fig. 9 and 10). This can be verified by the EMG measurements of the RF and BF muscles. In the time interval $[55,90]$s, the RF muscle is active during the flexion phase of the exoskeleton and the BF muscle is active during the extension phase. Notice again the time lag between the current and desired knee joint angles since it is the classical $\mathcal{L}_1$ adaptive control law that is implemented. Note that the control torque is greater when the wearer is resisting the exoskeleton’s movement to overcome this resisting muscular effort and to drive the shank-foot-exoskeleton system towards the desired trajectory. The wearer’s resistance has allowed the deviation of the current trajectory from the desired one (first plot of Fig. 9).

The resistance of the wearer can also be seen in the estimated parameters plots through the appearance of some oscillations (cf. Fig. 10).

In the last experiment, the augmented $\mathcal{L}_1$ adaptive control is implemented with a passive phase in the time interval $[0,49]$s followed by a resisting phase in the time interval $[49,90]$s (cf. Figs. 11 and 12). This can be verified by the EMG measurements of RF and BF muscles. One can notice that the time lag in the knee joint angle tracking has been eliminated. Note that, as in the previous experiment, the control torque has higher values during the resisting phase relative to the passive one. Note also that during the resistive phase, the current knee joint angle does not track accurately the desired one. The estimated parameters (cf. Fig. 12) present less oscillations relative to the classical $\mathcal{L}_1$ adaptive control (cf. Fig. 10).
V. CONCLUSION AND FUTURE WORK

In the present paper, a classical $L_1$ adaptive control is applied to control the lower limb exoskeleton EICOSI. Because of the time lag that this control introduces, an augmented $L_1$ adaptive control by means of a nonlinear proportional controller is proposed. These two control laws are tested in real-time using the exoskeleton EICOSI for assistance-as-needed and resistive rehabilitation case studies while tracking a desired trajectory considered defined by a therapeutic doctor. Experiments have confirmed the presence of a time lag when the classical $L_1$ adaptive control is applied. This time lag has been eliminated when the augmented version of the $L_1$ adaptive control is applied. When performing assistance-as-needed experiments, results have showed that the control torque has a lower value relative to the passive case since the wearer is performing the movement and the exoskeleton is only assisting him. On the other hand, when performing the resistive rehabilitation, the control torque has higher values than the passive case in order to counteract the resistive effort of the wearer and drive the exoskeleton towards the desired trajectory.

Future works consist in applying the $L_1$ adaptive control on the full lower limb exoskeleton for achieving daily living activities like walking, ascending and descending stairs.

REFERENCES


