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ACTIVE KNEE ORTHOSIS ADAPTIVE IMPEDANCE CONTROL BASED ON THE HUMAN IMPEDANCE DETERMINED BY POLYNOMIAL METHOD

Denis Mosconi
Adriano A. G. Siqueira
University of São Paulo, São Carlos School of Engineering
Av. Trabalhador São-Carlense, 400, São Carlos-SP, 13566-590, Brazil
denis.mosconi@ifsp.edu.br, siqueira@sc.usp.br

Abstract. *The development of interaction controls is of extreme importance for the field of robot assisted therapy, since such controls must meet the safety and efficacy requirements of the treatment, guaranteeing the good and safe recovery of the patient. In this work we present and simulate an active knee orthosis adaptive impedance control whose adaptation law is based on the human knee impedance and whose purpose is to help the patient to perform the rehabilitation exercises, providing an auxiliary torque that ensures the maintenance of the amplitude and pattern of movement. To determine the human knee impedance, we also propose a novel polynomial method, seeking to introduce a tool that is able to determine human impedance directly from EMG signals, without the need of to estimate the human torque. Simulations of the proposed control were carried out using a human-orthosis computational interaction model and experimental data of angular position and EMG signals from a user performing sinusoidal movements of extension and flexion of the knee. When compared to the traditional non-adaptive impedance control, the adaptive impedance control showed a better response, however it has the limitation of computational costs involved in determine the user impedance. The polynomial method proved to be feasible but not useful, since it requires fine adjustments for each rehabilitation exercise, which is undesirable and often not viable.*

Keywords: *Human-orthosis interaction model, OpenSim, Interaction Control.*

1. INTRODUCTION

Robotic neurorehabilitation is a powerful resource in the motor recovery of stroke victims. Several types of robots have been developed to ensure effectiveness and meet the growing demand for this type of treatment (Vinoj *et al.*, 2019; dos Santos *et al.*, 2017b; Young and Ferris, 2017; Huo *et al.*, 2016), since, according to the World Health Organization, in 2030 approximately 200 million people around the world will have some motor dysfunction due to stroke (WHO, 2011, 2018).

Ensuring a good interaction between the patient and the robot is extremely important to achieve an effective and safe treatment. For this purpose, interaction controls have been widely studied (Peña *et al.*, 2019; Pérez-Ibarra *et al.*, 2019; Khamar and Edrisi, 2018; Nunes *et al.*, 2018; Jutinico *et al.*, 2017), great part of them based on the impedance control proposed by Hogan (1985).

The purpose of this work is to present and simulate an active knee orthosis adaptive impedance control whose adaptation law is based on the human knee impedance and whose purpose is to help the patient to perform the rehabilitation exercises, providing an auxiliary torque in order to ensure the maintenance of the amplitude and pattern of movement. To determine the human knee impedance, we also propose a novel polynomial method, seeking to introduce a tool that is able to determine human impedance directly from EMG signals, without the need of to estimate the human torque which could to introduce uncertainties and to increase the computational cost.

We propose two hypothesis: (1) the polynomial method proposed is a useful resource to determine the human impedance through the muscle activations. (2) the proposed adaptive control has a better effect than traditional non-adaptive control to help the patient to perform a desired movement.

The efforts employed in this work are justified by the need to seek new methods with low computational cost in determine the user impedance through EMG signals, as well as, to provide interaction controls that are connected with the patient not only through his kinematics, but also through his biomechanical aspects.

2. METHODOLOGY

The methodology of this work can be divided into five parts: experimental procedure, determination of the human knee impedance, proposition of the adaptive impedance control, simulation and analysis of the results. Each of these parts are presented in detail below.

2.1 Experimental Procedure

The experimental procedure consists of a human wearing an active knee orthosis and, in a seated position, performing sinusoidal movements of flexion and extension of the knee. Through a graphical user interface, the subject can see the desired trajectory and the current position of the set knee-orthosis, so that the user is asked to follow the reference as closely as possible.

While the subject performed the knee movement, a Trigno™Wireless EMG system (Delsys Inc., Natick, MA, USA) was used to measure the EMG signals of five muscles involved in the movement: the extensors rectus femoris (RF), vastus medialis (VM) and vastus lateralis (VL), and the flexors biceps femoris (BF) and semitendinosus (SM).

The active knee orthosis used in this work is one of the six free and independent joints from ExoTAO: a modular lower limb exoskeleton developed by (dos Santos *et al.*, 2017b), that performs movements in the sagittal plane (Fig. 1). Such knee orthosis is equipped with a series elastic actuator (SEA) designed by (dos Santos *et al.*, 2017a) that was used to measure the knee angular position and the torque applied by the actuator of the orthosis.

Figure 1. Active knee orthosis from ExoTAO

During the experimental procedure the active orthosis was drove by an non-adaptive impedance control described by Eq. (1)

$$\tau_{non-adaptive} = K_r(\theta^d - \theta) - B_r\dot{\theta} \tag{1}$$

Where $K_r = 60N.m/rad$ is the stiffness coefficient and B_r is the damping parameter.

The experimental procedure was performed by a health male 29-year-old subject with 1.77 m height and 84 kg mass. Its anthropometric data were collected and used to fit a human-orthosis interaction computational model in order to carry out the simulations.

2.2 Human Knee Impedance

In this work we propose a method to determine the mechanical impedance of the human knee based on the muscular activations determined from the EMG signals measured. Such activations were obtained as described by (Peña *et al.*, 2019).

To determine the human knee impedance, a 9th order polynomial is proposed, as can be seen in the Eq. (2). In this case, $m = 5$ refers to the number of muscles whose activations were determined and $n = 9$ refers to the order of the polynomial. The activations are $x^j_{(i,j)}(t)$ and $a_{(i,j)}$ are the coefficients of the polynomial. a_0 is a constant term.

$$K_{user}(t) = a_0 + \sum_{i=1}^{m}\sum_{j=1}^{n} a_{(i,j)}x^j_{(i,j)}(t) \tag{2}$$

The coefficients of the polynomial (a_0 and all $a_{(i,j)}$) were obtained by the least-square method, according to the Eq. (3).

$$a = (A^T A)^{-1} A^T b \tag{3}$$

Where a_{46x1} is a vector of the polynomial coefficients, A_{500x46} is a matrix of the activations and its exponentiation and b_{500x1} is a vector of the human knee impedance determined by Peña *et al.* (2019) that was considered a reference for this work.

The intention in proposing such a method is to evaluate a possible alternative to determine the human knee impedance through a shortest path. The method used by (Peña *et al.*, 2019) requires first the determination of the human torque, which is achieved by means of muscle and joint modeling, such method requires more computational effort and involves more parameters of uncertainty (Fig. 2b). We intend to evaluate the possibility of having a direct process between muscle activations and impedance (Fig. 2a).

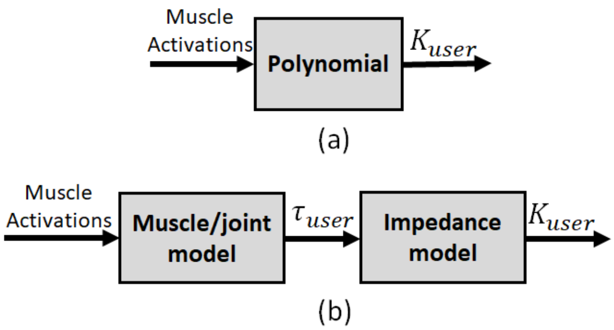


Figure 2. Comparison between the flowchart of the proposed method (a) and the method used by Peña *et al.* (2019) (b) to determine the human knee impedance

The determination of the polynomial coefficients was made in two steps, using two set of data obtained from the experimental procedure: the first step refers to the determination of the coefficients properly and the second one refers to the validation of these coefficients in order to verify the effectiveness of the polynomial.

2.3 Adaptive Impedance Control

We propose an adaptive impedance control to the active knee orthosis, in such a way that the virtual stiffness of this control is adapted according to the user stiffness, the torque required to complete the movement and the tracking error of the desired trajectory. Equation (4) expresses the control law.

$$\tau_{adaptive} = K_r^{adap}(\theta^d - \theta) - B_r\dot{\theta} \tag{4}$$

Where $\tau_{adaptive}$ is the orthosis torque, θ^d is the desired angular position of the knee, θ is the real position, $\dot{\theta}$ is the angular velocity, $B_r = 0.5\ N.m.s/rad$ is the virtual damping parameter and K_r^{adap} is the virtual stiffness coefficient which is iteratively adapted according to Eq. (5).

$$K_{r(k)}^{adap} = fK_{r(k-1)}^{adap} + (1 - f)\overline{K} \tag{5}$$

Being $f = 0.85$ a forgetfulness factor and $\overline{K}$ determined by Eq. (6).

$$\overline{K} = min(\alpha\frac{\tau_{nec}}{\theta_e}, \frac{1}{\beta K_{user}}) \tag{6}$$

Where τ_{nec} is the torque necessary to accomplish the movement (determined through *Inverse Dynamics Tool* from OpenSim), $\theta_e = \theta^d - \theta$ is the position error, K_{user} is the user impedance, determined through the polynomial presented in Section 2.2, $\alpha = 1$ and $\beta = 0.5$ are the gains.

2.4 Simulation

In order to verify the feasibility of the human impedance determination method and the adaptive control proposed, we carried out simulations using a human-orthosis computational interaction model. Such interaction model is composed by a 3D lower limbs neuromusculoskeletal model (*gait2392*) provided by OpenSim[1] (Delp *et al.*, 2007) in whose knee articulation were placed coordinate actuators to represent the orthosis.

The anthropometric data from the subject that performed the experimental procedure were used together the *Scale Tool* from OpenSim in order to fit the computational model to the human and thus obtain simulations with output values comparable to the real ones.

The simulations were carried out in a forward dynamics-base algorithm developed in MATLAB® (Fig. 3). Such algorithm applies the torques from the controls (in this case the human control that represents the human behavior and the robot control that represents the impedance controls tested) to the interaction model and then, through numerical integration determines the kinematic quantities (e.g. knee angular position and velocity).

In the Fig. 3 a flowchart of the simulation algorithm is presented. The **Polynomial** block contains the polynomial proposed in Section 2.2 that determines the human impedance K_{user}. The **Robot Control** block contains the impedance control of the orthosis (adaptive or non-adaptive, according to the simulation performed) that determines torques applied by the orthosis (τ_{exo}) and the **Human Control** contains the *Inverse Dynamics Tool* from OpenSim, that determines the user torque (τ_{user}). The angular position reference from the experimental procedure (θ^d) was used and compared to the angular position obtained with the simulations ($\hat{\theta}$).

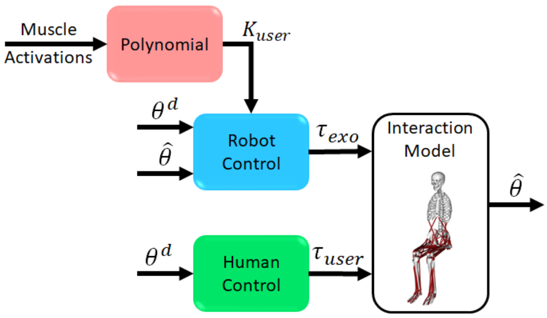


Figure 3. Flowchart of the algorithm used in the simulations.

Both the human-exoskeleton interaction model and the forward dynamics-based algorithm were developed and validated in previous work (Mosconi *et al.*, 2019). In the present work we are interested in use them as a tool to simulate the impedance control.

All simulations were carried out on a computer with Intel®Core™i7-5500 2.40 GHz processor, 8.00 GB of RAM, 2.00 GB dedicated video card, Windows 10 Home Single Language 64 bits. The OpenSim version 3.3 and the MATLAB R2017b were the platforms where the simulations took place.

2.5 Analysis

The effectiveness of the human impedance determination method proposed was evaluated comparing the results obtained with the polynomial with the ones obtained by (Peña *et al.*, 2019) (whose results have already been validated). It is expected that the results obtained by the method proposed in this work are as close as possible to the method used by Peña *et al.* (2019), so our method can be considered useful.

With regard to adaptive impedance control, it is expected that the torques obtained through such control are capable of making the interaction model perform the same movement described by the position reference (θ^d) during the experimental procedure, that is: we expected $\hat{\theta} \approx \theta^d$. To verify this results, we made an analysis comparing the knee angular position obtained with the simulations (adaptive impedance control) and the ones obtained through the experimental procedure (non-adaptive impedance control).

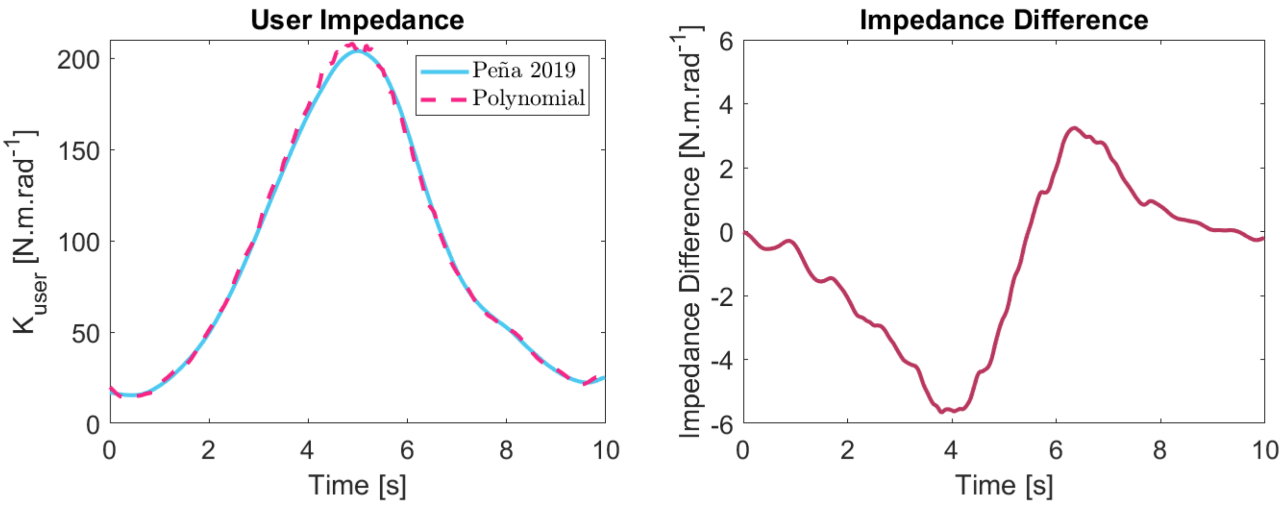
3. RESULTS

A comparison between the human knee impedance obtained with the polynomial method proposed and the one obtained by Peña *et al.* (2019) is presented in Fig.4 (a). It is possible to notice that the values obtained with the proposed method are close to the ones obtained by the method used by Peña *et al.* (2019), which prove that the polynomial is a feasible to determine the impedance having the muscular activations.

[1]http://opensim.stanford.edu

Looking at Fig. 4 (b), it is possible to notice that the difference between the impedances obtained by the polynomial method and that of Peña *et al.* (2019) is small, showing the capacity of the polynomial.

However, it was noted during the tests that, with different sets of data, the polynomial needed fine adjustments to correctly determine the human impedance with an error of less than 5%. Such fine adjustments are undesirable when using the impedance obtained to perform controls (such as the proposed adaptive impedance control), so, although it is feasible, the polynomial method may not be the best method for determining human impedance.

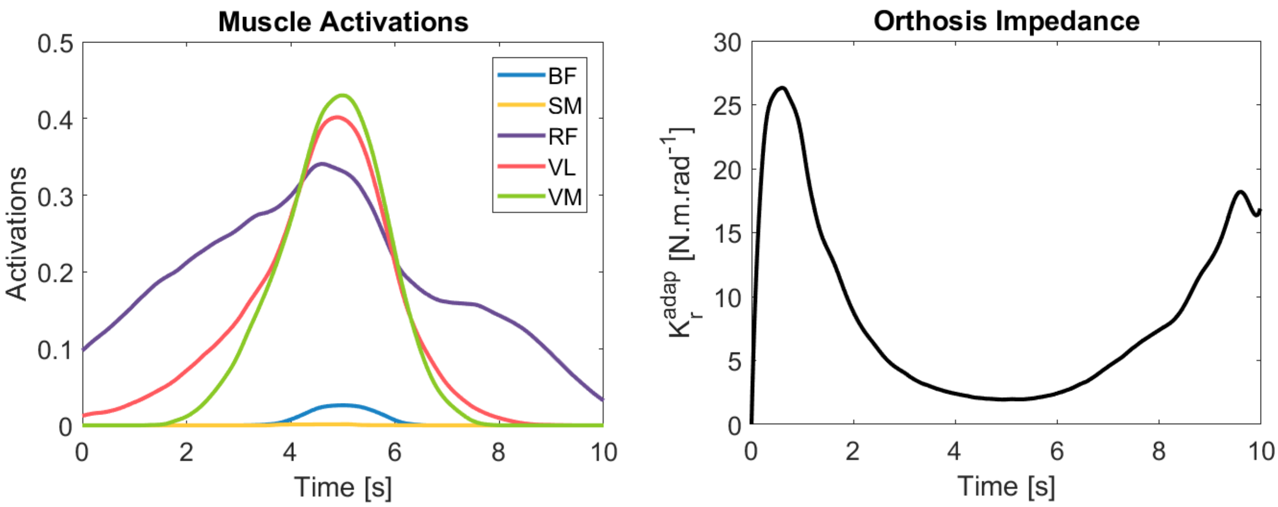


(a) (b)

Figure 4. (a) Comparison between the impedance obtained through the proposed method (Polynomial) and the one obtained through the method proposed by Peña *et al.* (2019), (b) difference between the human impedances obtained from the different methods presented

The muscle activations obtained during the experimental procedure are depicted in the Fig. 5 (a). The muscles involved are: biceps femoris (BF), rectus femoris (RF), semitendinosus (SM), vastus intermedius (VI), vastus lateralis (VL) and vastus medialis (VM). These activations were used to determine the human knee impedance (K_{user}) through the polynomial method proposed.

The variable impedance of the orthosis, obtained from Eq. (5)-(6) is presented in Fig. 5 (b). It is possible to notice that when the user impedance is high (see the instant 5s in Fig.4 (a)) the orthosis impedance is low, attempting the assist-as-needed protocol and helping the user only when necessary (what is perceived through its low impedance).

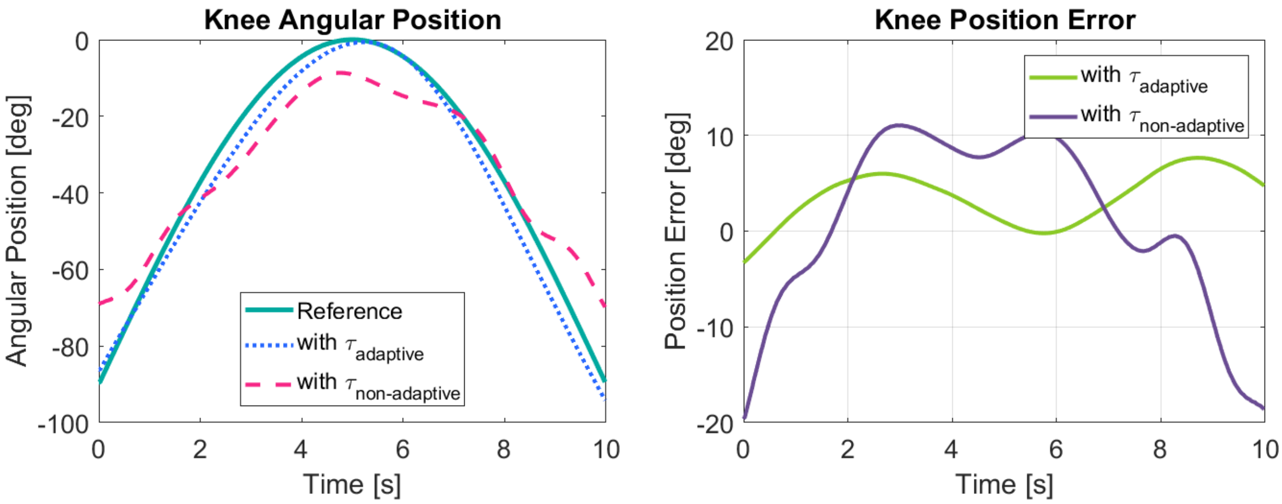


(a) (b)

Figure 5. (a) Muscle activations measured during the experimental procedure, (b) the adapted impedance of the orthosis

A simulation using the orthosis torque measured during the experimental procedure ($\tau_{non-adaptive}$ applied according to the control law expressed by equation (1)) and one simulation using the orthosis torque obtained from the proposed adaptive impedance control ($\tau_{adaptive}$, equation (4)) were performed. A comparison between the knee angular position obtained with these torques and the reference are presented in Fig. 6 (a). It is possible to notice that with the orthosis controlled by the adaptive impedance control, the user followed the reference much better than in relation when using the orthosis controlled by the non-adaptive control. In addition, the adaptive control ensured a movement with greater amplitude, which is beneficial for the articulations of the human being, as well as maintaining the health of the muscle-tendon set.

A comparison between the errors of following the trajectory with $\tau_{non-adaptive}$ and $\tau_{adaptive}$ is shown in Fig. 6 (b). With $\tau_{non-adaptive}$ the error was greater while with $\tau_{adaptive}$ the error remained within a limit range, with a small oscillation.

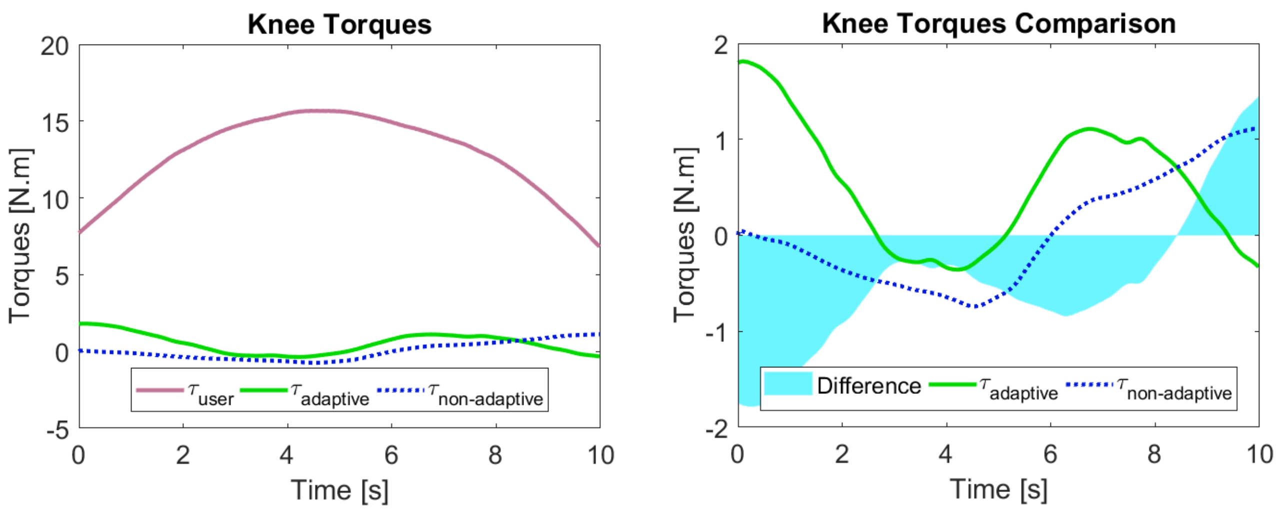


(a) (b)

Figure 6. (a) Comparison between the angular position of the knee when the torque obtained from the adaptive impedance control and the one obtained from non-adaptive impedance control are applied to the exoskeleton, (b) angular position error of the knee

The human torque (τ_{user}) and the orthosis torques from the adaptive and non-adaptive controls ($\tau_{adaptive}$ and $\tau_{non-adaptive}$, respectively) are presented in Fig. 7 (a). As the user is a health subject, the torques applied by the orthosis are small when compared to the user. It is evident that the torque the user performed has the same shape as the movement performed.

The Figure 7(b) presents a comparison between the $\tau_{adaptive}$ and $\tau_{non-adaptive}$, showing the difference between them. It is possible to verify that the $\tau_{adaptive}$ is, in module, greater than the $\tau_{non-adaptive}$, a fact that ensured the good tracking of the desired trajectory.



(a) (b)

Figure 7. (a) Torques on the knee. The τ_{user} corresponds to the torque performed by the subject, (b) comparison between the torques obtained from the adaptive and non-adaptive impedance controls

For a third simulation, a PID control feedback loop was added to human control (Fig. 8). Whit this, we approximate to the feedforward-feedback human control concept ((Gillespie *et al.*, 2016; Frey *et al.*, 2011; Wolpert *et al.*, 1998)) and seeks to reduce the tracking errors. The PID gains were determined by trial and error and its values are: $Kp = 6.3$, $Ki = 5.8$ and $Kd = 5.5$.

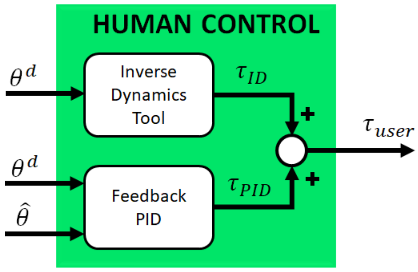


Figure 8. Human control based on the feedforward-feedback concept.

Observing the Fig. 9 (a) it is possible to notice that the tracking errors were reduced, and the movement was carried out more smoothly, without the oscillations observed when the non-adaptive impedance control is used. The tracking error is smaller when compared to the control without PID and, like the movement execution, it does not show relevant oscillations Fig. 9 (b).

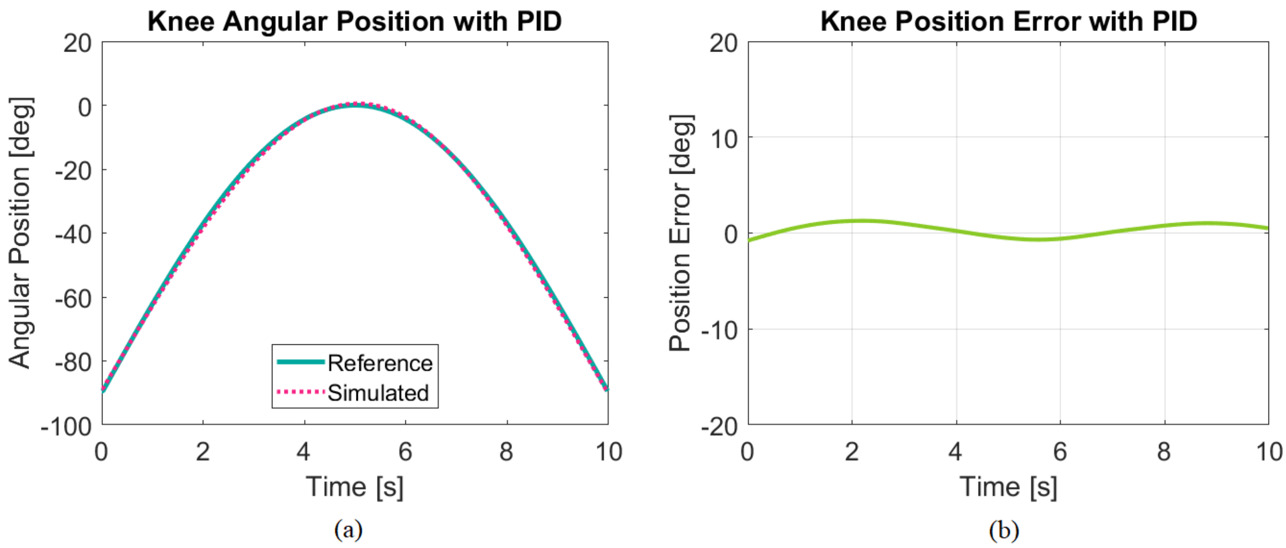


Figure 9. (a) Knee angular position when a PID loop was included to the human control, (b) knee angular position error when a PID loop was included to the human control

The torques involved in this simulation are presented in the Fig. 10. In this case, the τ_{ID} has the same value as the τ_{user} in the previous simulations and now $\tau_{user} = \tau_{ID} + \tau_{PID}$. The τ_{PID} is small, as there was little trajectory tracking error to be corrected (see Fig. 6 (b) - $\tau_{adaptive}$).

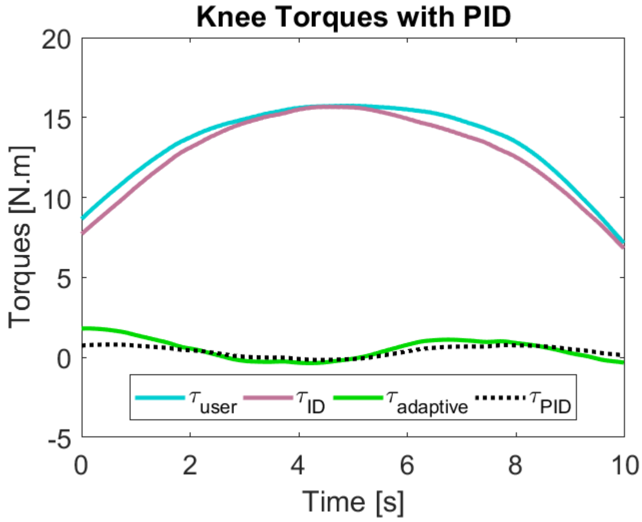


Figure 10. Torque on the knee when a PID loop was included to the human control

A comparison between the RMS and the maximum angular position error obtained from the simulations is presented in Tab. 1. In relation to the non-adaptive control, the adaptive control had a lower RMS error but a greater maximum error. When the feedback PID was included to the human control, both the RMS and maximum errors decreased, being the lower between all simulations.

Table 1. Comparison between the RMS and maximum angular position error with the controls studied.

Control	Error RMS	Error Maximum
Adaptive	2.29	5.52
Non-adaptive	2.42	4.28
Adaptive and Human with PID	0.82	1.55

All values are in degrees.

Regarding the hypotheses presented, the following can be said:

- **Hypothesis 1:** The polynomial method is a possible tool to determine the human knee impedance through the muscle activations, however, it is not useful since it needs fine adjustment for each experimental procedure, what is undesirable.

- **Hypothesis 2:** The adaptive impedance control proposed is profitable, and reasonably better than the non-adaptive one, however, the computational cost in determining the human impedance necessary for such control can give it limitations in certain applications.

4. CONCLUSIONS

In this work we presented a polynomial method to determine the human knee impedance based on the muscle activations and tested it with an adaptive impedance control also proposed by us.

The polynomial method proved to be feasible but not useful, since it requires fine adjustments for each rehabilitation exercise, which is undesirable and often not viable.

The adaptive impedance control showed good results when compared to the traditional non-adaptive impedance control, however this control has the limitation of computational costs involved in determine the user impedance.

For future work we intend to improve and test the adaptive impedance control, applying it experimentally to the active knee orthosis based on ExoTAO and used in rehabilitation therapy exercises.

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