Hybrid Impedance and Nonlinear Adaptive Control for a 7-DoF Upper Limb Rehabilitation Robot: Formulation and Stability Analysis

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Abstract: Physical rehabilitation aims to improve the condition of people with any musculoskeletal disorder. Different assistive technologies have been developed to provide support to this process. In this context, human-machine interaction has progressively improved to avoid abrupt movements and vibrations, to obtain a more natural interaction, where control strategies play a key role. In this work, a control technique based on the combination of nonlinear adaptive theory with a hybrid impedance control applied to a 7-DoF upper limb assistance robotic device is proposed. Additionally, we include the stability analysis using Lyapunov functions. Then, we validate the strategies through simulations for one rehabilitation routine test. The articular and cartesian obtained results demonstrate the effectiveness of the control to follow trajectories. The control stabilizes the trajectories in 0.9 seconds even when the initial conditions start far from the desired trajectories, without producing vibrations or overshoots, which is the desired behavior in rehabilitation applications like the one we propose.

1 INTRODUCTION

People who suffer any trauma or musculoskeletal disorder require a rehabilitation process that focuses on improving the functional capabilities and allows the patient to recover socially, physically, and occupationally (Ritchie, 2003; hil, 2012; reh, 1950). To regain the limb functionality, the patients undergo treatments that include exposing the muscular tissues to stress in a progressive and appropriate manner, increasing the range of mobility and muscle strength progressively (McHugh et al., 2013; Wattchow et al., 2018; Bruder et al., 2017; Milicin and Sîrbu, 2018; Gates et al., 2015; Ritchie, 2003).

Assistive technologies are oriented to support physical rehabilitation processes, adapting to the patients' condition, according to their disability (Linda et al., 2018; Olanrewaju et al., 2015). The use of these technologies has increased in the last few years due to the effectiveness of the therapy (Ballantyne and Rea, 2019; Molteni et al., 2018). Moreover, assistive technologies allow the acquisition of accurate measurements through sensors or smart devices to evaluate the progress of the patients (Munih and Bajd, 2011). Rehabilitation robots may be used for recovery or compensatory purposes to improve the rehabilitation processes and the patients' quality of life in the shortest possible time (reh, 1950).

Rehabilitation robots are designed to adapt to the conditions of motion and strength assistance, according to the level of intervention that the patient requires. According to (Mancisidor et al., 2019a; Akdoğan et al., 2018) we define five levels of assistance, namely (i) passive, that requires total robot intervention, (ii) assistive, that requires partial robot intervention, (iii) isotonic, which means no robot intervention (Munih and Bajd, 2011; Trochimczuk et al., 2018; Akdoğan et al., 2018), (iv) isometric, where there is a robotic-supplied static muscle level contraction, and (v) resistive, where there is a robotic-supplied dynamic muscle strengthening (Munih and Bajd, 2011). The assistance modes allow the parameterization of the exercises according to the patients' condition. Here is where a proper control strategy comes into play (Meng et al., 2015).

Rehabilitation robots are systems based on human-machine interaction at the clinical level. In this interaction, an adequate control strategy is neces-

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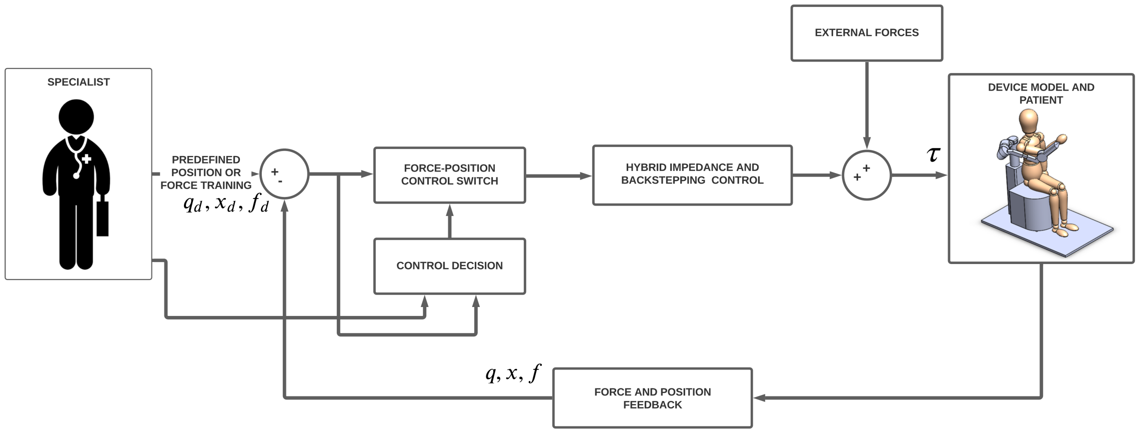


Figure 1: General scheme of proposed control.

sary to prevent further harm to the user. For this, the controller have to accurately track trajectories within the rehabilitation routines, and compensate the system against disturbances and unwanted loads (phy, 1932).

Control strategies for rehabilitation applications are designed to improve the system's performance and compensate undesired behaviors that may compromise the integrity of the person (Marchal-Crespo and Reinkensmeyer, 2009; Meng et al., 2015). In the literature, control strategies are designed to improve the interaction capabilities of robotic systems with patients. This implies that both the rehabilitation systems and the control strategies have a certain level of complexity in the design, especially if they are used in robotic systems of more than 6-DoF.

In this paper we propose a control strategy that combines hybrid impedance and nonlinear adaptive control for assistive rehabilitation robotic systems.

In the literature, we find recent works that propose control strategies applied to assistive robotic systems, focused on performing more natural motions. We have evidenced that one of the most used controls is impedance control. This strategy tries to imitate natural movements taking as a reference a desired damper-spring-mass model. The controller can be force-based or position-based (Jutinico et al., 2017; Song et al., 2017). In this type of control, a concept called Assist as needed (AAN) has been implemented. This is an assistance strategy that relates directly the levels of intervention that the robot provides to the patient. For example, in (Wu and Wu, 2018) an impedance control, a.k.a multimodal control is applied to a therapeutic exoskeleton for upper limb rehabilitation. In (Asl et al., 2020) a control strategy is developed to maximize the patient's participation in the rehabilitation process, using the AAN strategy with an impedance controller for speed tracking, and a velocity tracking controller, which adjusts its contribution in an AAN way, by monitoring the tracking error. In the same way, in (Chen et al., 2016) a control strategy based on ANN is applied to a 7-degrees-of-freedom (DoF) upper limb rehabilitation system. Moreover, in (Mancisidor et al., 2019b), an inclusive control based on adaptive assist modes is applied for upper limb rehabilitation using transition impedance control based on position and strength. It is worth to remark that the stability of impedance controllers is usually analyzed by theoretical methods such as Lyapunov equations, which are constrained to design conditions and workspaces according to the application. In general, to the best of our knowledge we have not found in the literature an impedance controller that can adapt to the dynamics of a system, and to other dynamics and uncertainties. In this work, we propose the adaptability that is required.

On the other hand, hybrid strategies have been proposed to enhance the capabilities of impedance controllers, see for instance, (Akdoğan et al., 2018). In the same way, an adaptive impedance control based on backstepping theory and fuzzy logic has been presented in (Bai et al., 2019). In (Li et al., 2017), an adaptive impedance control is presented using electromyography signals (sEMG) that are used to design the optimal reference impedance model. Furthermore, an adaptive neural network control with a high-gain observer is developed to approximate the effect of the dead zone and the robot's dynamics. Like impedance control, stability in this type of controls plays a very important role and must to be analyzed.

In the same way, alternatives of controls have been proposed, some that use EMG as the main basis such as (Lee et al., 2017) where one proposes a novel control method to minimize muscle energy for robotic systems that support the movements of a user under unknown external disturbances, using sEMG. Others that use alternatives for the integration of control methods such as (Brahmi et al., 2018) where it proposes an adaptive control through rehabilitation tasks using integrated backstepping theory with time-delay estimation, or as in (Miao et al., 2020) where a new strategy of control for a bilateral upper limb system using regulation based on human-robot interaction force measured using compliance control based on position controller (low level) and admittance controller (high level). Also in (Huang et al., 2018), an adaptive control based on force is proposed under the AAN concept. Or also as an alternative in (Wu et al., 2018), where a neural-diffuse adaptive controller (NFAC) based on a radial basis function network (RBFN) is proposed to guarantee the precision of the path tracking with parametric uncertainties and environmental perturbations of an upper limb exoskeleton. All these proposed techniques generate new alternatives to improve or propose alternative or complementary control strategies on this type of applications in physical rehabilitation.

In this paper, we design a control technique based on the combination of nonlinear adaptive theory and hybrid impedance control (position-based and force-based strategy) applied to a 7-DoF upper limb assistive robotic system. This is a interesting strategy that exploits the characteristics of both techniques. We carry out a stability analysis using Lyapunov functions to prove the performance of the control, considering the stabilization time, the response to disturbances, and the error tracking. This work is a previous step to the implementation in the physical assistive system. The control model that we propose aims to improve and widen the application of (impedance) nonlinear adaptive control strategies to assistive rehabilitation systems. The proposed control model is shown in Fig. 1. It consists in an adaptive nonlinear control using an impedance control strategy, with the option of switching between position-based or force-based control when required, for a 7-DoF assistive rehabilitation device. The results obtained from the simulation of three common routines are the positions of the trajectories in the joint space and the positions and orientations of the end-effector trajectories. The controller tracks with a maximum error of 2% the desired trajectories. These results show the effectiveness of the strategy proposed.

This paper is organized as follows: section II presents the proposed assistive robotic system and the dynamic formulation. In Section III we describe the proposed control strategy and the stability analysis performed using Lyapunov functions. The set up for the validation and simulation results of common routine trajectories in upper limb rehabilitation are presented in Section IV. Finally we give some conclusions and future work perspectives in Section V.

2 ASSISTIVE ROBOTIC DYNAMIC MODEL

We have designed a 7-Dof assistive robotic system for the rehabilitation of upper limb tendinopathies. This system allows the main movements of each joint of the upper limb, which are: (i) scapular band, protraction/retraction movements. (ii) shoulder, flexion/extension, abduction/adduction and internal/external rotation movements. (iii) elbow, flexion/extension movements. and (iv) wrist, flexion/extension, and pronation/supination movements. Fig. 2 shows the concept of the robotic system and the simplified diagram for the calculation of the transformation matrices, using the Denavit-Hartenberg convention. Table 1 shows the corresponding parameters to obtain the transformation matrices and the Jacobian $J(q)$ for the robot's dynamic model and control formulation.

Table 1: Denavit-Hartenberg parameters.

$Joint_i$	θ_i	d_i	a_i	α_i
1	θ_1^*	L_1	L_3	0
2	$\theta_2^* - \pi/2$	L_4	0	$-\pi/2$
3	$\theta_3^* - \pi/2$	L_5	0	$-\pi/2$
4	θ_4^*	0	$-L_6$	$\pi/2$
5	θ_5^*	0	0	$\pi/2$
6	$\theta_6^* + \pi/2$	L_7	0	$\pi/2$
7	$\theta_7^* + \pi/2$	0	L_c	0

Let us write the general dynamics model of the assistive device in the joint space as,

$$M(q)\ddot{q} + C(q,\dot{q})\dot{q} + G(q) = \tau_{total} \tag{1}$$

Where q is the vector R^{7x1} of generalized joint coordinates, $M(q)$ is the inertia matrix R^{7x7}, $C(q,\dot{q})$ is the Coriolis matrix R^{7x7}, $G(q)$ is the vector of gravity forces R^{7x1} (due to the weight of each link) and τ : is the vector of generalized (non-conservative) forces R^{7x1}.

Through the Euler-Lagrange formulation, the Lagrangian operator is

$$L(q,\dot{q}) = K(q,\dot{q}) - P(q) \tag{2}$$

Where $K(q,\dot{q})$ is the kinetic energy and $P(q)$ is the potential energy of the system, then

$$K(q,\dot{q}) = \frac{1}{2}\dot{q}^T M(q)\dot{q} \tag{3}$$

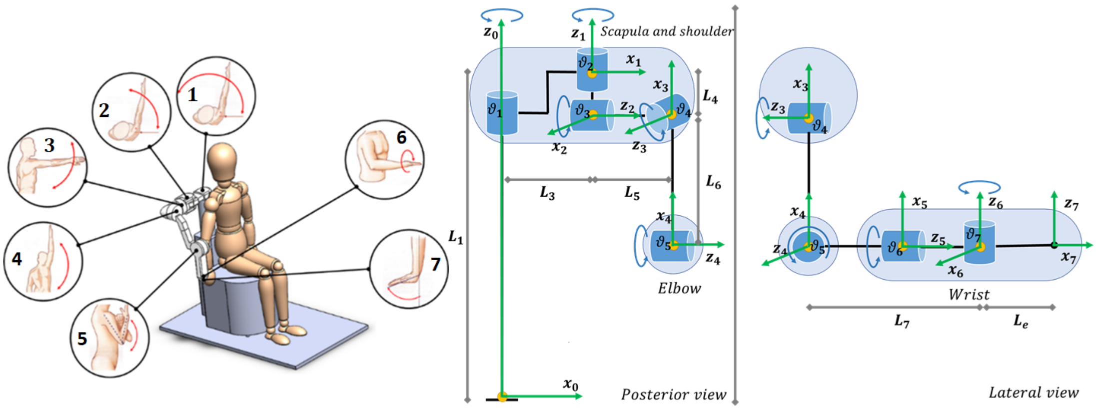


Figure 2: Concept design and simplified diagram of the robotic system. Movements performed: 1. protraction/retraction. 2. internal/external rotation. 3. flexion/extension. 4. abduction/adduction. 5. flexion/extension. 6. pronation/supination and 7. flexion/extension.

and

$$M(q) = \sum_{i=1}^{n} (m_i J_v^{iT} J_v^i + J_\omega^{iT} R_i \tilde{I}_i R_i^T J_\omega^i) \tag{4}$$

Where m_i is the mass of link i, v_{ci} is the velocity of the center mass of link i, ω_i is the angular velocity of the center of mass of link i $\tilde{I}_i$ is the constant inertia tensor, $J_v^i(q)$ is the lineal velocity component of the Jacobian matrix of link i and $J_\omega^i(q)$ is is the angular velocity component of the Jacobian matrix of link i. Then, the potential energy is defined as

$$P(q) = -\sum_{i=1}^{n} P_{ci} m_i g_0^T \tag{5}$$

Where m_i is the mass of link i, g_0 is the gravity vector and P_{ci} is the position of the center of mass of link i. Considering $J_v^i(q) = \frac{\partial P_{ci}}{\partial q}$, the general vector of gravity forces $G(q)$ is

$$G(q) = -\sum_{i=1}^{n} J_v^{iT}(q) m_i g_0 \tag{6}$$

$G_i(q)$ is the moment of joint i due to the gravity (weight). Then, the coriolis matrix $C(q,\dot{q})$, can be defined by means of Euler-Lagrange formulation, using the Christoffel symbols of the first kind. Then, we have

$$c_{ij} = \sum_{k=1}^{n} c_{ijk} \dot{q}_k \tag{7}$$

$$c_{ijk} = \frac{1}{2}\left(\frac{\partial m_{ij}}{\partial q_k} + \frac{\partial m_{ik}}{\partial q_j} - \frac{\partial m_{jk}}{\partial q_i}\right) \qquad c_{ijk} = c_{ikj}$$

Where n is the number of DoF, m_{ii} is the moment of inertia at the $i-th$ joint, when the other joints do not move. m_{ij} is the $i-th$ joint acceleration effect at the joint j (coupling effect), $c_{ijj}\dot{q_j}^2$ is the centrifugal force at joint i due to the $j-th$ joint velocity and $c_{ijk}\dot{q}_j\dot{q}_k$ is the Coriolis effect on $i-th$ joint, due to $j-th$ and $k-th$. joint velocity

Let us define the components of the generalized forces τ_{total} as:

$$\tau_{total} = \tau - f_{ext} - f(q,\dot{q}) \tag{8}$$

Where τ is the torque applied (by the actuators) to the joints, f_{ext} is the torque due to external forces/moments and $f(q,\dot{q})$ is the torque due to friction in the joints defined as

$$f(q,\dot{q}) = F_s sgn(\dot{q}) + F_v \dot{q} \tag{9}$$

Where F_s is the static friction matrix and F_v is the viscous friction matrix (Akdoğan et al., 2018; Brahmi et al., 2018).

3 CONTROL STRATEGY: HYBRID IMPEDANCE AND NONLINEAR ADAPTIVE CONTROL

Impedance control method is widely used in rehabilitation systems. In this paper, hybrid impedance control and nonlinear adaptive control are combined into a single structure for a 7-DoF upper limb rehabilitation system. We propose a control scheme as shown in Fig. 1. Therefore, the desired mechanical impedance of the robot's end-effector can be adjusted while the robot follows the trajectory of the desired force or position. This behavior aims to reproduce the actions that the specialist performs during physiotherapy, regarding strength and position. In our case, a switching matrix allows the transition from the position-based control to the force-based control. This change and the level of intervention of the system depends on the therapist's decision or on the system's position or force error. Physical exercises are modeled using the hybrid impedance control approach and are classified considering the level of intervention. According to (Akdoğan et al., 2018), the desired hybrid impedance can be stated as

$$-J(q)^T f_{ext} = M_d(\ddot{x} - S\ddot{x_d}) + B_d(\dot{x} - S\dot{x_d}) + SK_d(x - Sx_d) + (I - S)F_d \tag{10}$$

Where M_d is the desired inertia matrix R^{6x6}, B_d s the desired damping matrix R^{6x6}, K_d is the matrix of desired stiffness R^{6x6}, x is the position and orientation vector of the end effector R^{6x1}, x_d is the desired position and orientation vector of the end effector R^{6x1}, F_d is the desired force vector R^{6x1} and S is a switching matrix between position-based and force-based controls, for more information about this control equation, we refer to (Akdoğan et al., 2018). Merging (1) and (10), we can write the hybrid impedance control law as

$$\tau = M(q)J(q)^{+}(S\ddot{x_d} + M_d^{-1}[(I-S)F_d - B_d(\dot{x} - S\dot{x_d}) - K_d(x - Sx_d) - J(q)^T f_{ext}] - \dot{J}(q)\dot{q}) + C(q,\dot{q}) + G(q) + f(q,\dot{q}) - f_{ext} \tag{11}$$

Where the velocity $\dot{x}$ and acceleration $\ddot{x}$ of the end effector are

$$\dot{x} = J(q)\dot{q} \quad \ddot{x} = J(q)\ddot{q} + \dot{J}(q)\dot{q}$$

Then, on the basis of the hybrid control definition, we reformulate and propose our control strategy. This formulation intends to implement hybrid impedance control with the demonstration of stability using nonlinear adaptive control theory by applying Lyapunov candidates, which guarantee stability, if there exists a function $V(x)$ that fulfills the following conditions:

- $V(x)$ positive definite for $x \neq 0 V(x) > 0$
- $\dot{V}(x)$ negative definite for $x \neq 0 \dot{V}(x) < 0$
- $V(x) \to \infty$ for $\|x\| \to \infty$
- $V(0) = 0$

Hybrid impedance control allows both position-based trajectory control when the patient needs to regain mobility, and force-based control when the patient needs to regain muscle strength. The disadvantage of using only this control is that there is no way to ensure stability when the position-force control transition is made. In this case, the nonlinear adaptive theory, with the Lyapunov candidates solves this drawback.

For the nonlinear adaptive formulation, let us define $\eta_1 = q$, $\dot{\eta_1} = \dot{q}$, $\eta_2 = \dot{q}$ and $\dot{\eta_2} = \ddot{q}$ and rewrite this terms into the general dynamic model equation (1). Merging (8) into (1) yields,

$$M(q)\dot{\eta_2} + C(q,\dot{q})\eta_2 + G(q) + f(q,\dot{q}) + f_{ext} = \tau \tag{12}$$

$$\begin{aligned}\dot{\eta_1} &= \eta_2 \\ \dot{\eta_2} &= M(q)^{-1}[\tau - C(q,\dot{q})\eta_2 - G(q) - f(q,\dot{q}) - f_{ext}] \\ &= U(t) - F(t)\end{aligned} \tag{13}$$

Where: $U(t) = M(q)^{-1}\tau$ and $F(t) = M(q)^{-1}[C(q,\dot{q})\eta_2 + G(q) + f(q,\dot{q}) + f_{ext}]$

Then by non-adaptive linear control methods, the control law can be stated as

$$\begin{aligned}U(t) &= F(t) + \dot{\xi} - K_2e_2 - e_1 \\ &= M(q)^{-1}[C(q,\dot{q})\eta_2 + G(q) + f(q,\dot{q}) + f_{ext}] + \dot{\xi} - K_2e_2 - e_1\end{aligned} \tag{14}$$

Where ξ is a virtual control input to $e_1 = \eta_1 - \eta_d$, η_d is the desired position and $e_2 = \eta_2 - \xi$. And consider that $\xi(t) = \dot{\eta_d} - K_1e_1$, where K_1 is a gain matrix of R^{7x7}, and K_2 is a gain matrix of R^{7x7}.

Then, the same procedure is applied to (10) of hybrid impedance rewriting with $e_s = (x - Sx_d)$, where e_s is the selective error that allows to switch between the expressions of position-based and force-based impedance control.

Now, let us introduce a variable as $\lambda_1 = e_s$, and the derivatives $\dot{\lambda_1} = \dot{e_s}$, $\lambda_2 = \dot{e_s}$ and $\dot{\lambda_2} = \ddot{e_s}$. Then we obtain the control law as

$$U_s(t) = F_s(t) - S\ddot{x_d} + \dot{\xi_s} - K_4e_4 - e_3 \tag{15}$$

Where $U_s(t) = -f_{ext} - (I - S)F_d$ y $F_s(t) = M_d^{-1}[B_d\lambda_2 + SK_d\lambda_1]$, $\xi_s(t) = S\dot{x_d} - K_3e_3$ is a second virtual control input, $e_3 = \lambda_1$, $e_4 = \lambda_2 = \lambda_2 + S\dot{x_d} - \xi_s$, K_3 is a gain matrix of R^{6x6} and K_4 is a gain matrix of R^{6x6} Finally, the control law that relates the system dynamics with the desired dynamics yields

$$U(t) = U_s(t) + [F(t) - F_s(t)] + [\dot{\xi} - \dot{\xi_s} + S\ddot{x_d}] + [K_4e_4 - K_2e_2] + [e_3 - e_1] \tag{16}$$

Notice that (16) defines the control system states as state errors. The errors compare the assistive system dynamics with the impedance desired dynamics. This control law requires the joint space errors, end-effector cartesian space errors and force errors. Therefore, we need to compute the kinematics and the Jacobian matrix to move from one space to the other during the simulation.

In addition, merging (14) into $\dot{V_1}$, and merging (16) into $\dot{V_2}$, we guarantee stability by proving Lyapunov functions are negative,

$$\begin{aligned}\dot{V_1} &= -e_1^TK_1e_1 - e_2^TK_2e_2 \\ \dot{V_2} &= -e_4^TK_4e_4\end{aligned} \tag{17}$$

Regardless the error, for $(K_1, K_2, K_4) > 0$, stability conditions of $\dot{V_1}$ and $\dot{V_2}$ will be satisfied.

4 CONFIGURATION AND SIMULATION RESULTS

For the validation of the control law in (16), we carried out three different tests where we define common rehabilitation routines. The tests consists in individual movements of each joint and combined joint movements of the whole upper limb. We defined the required joint trajectories for the following routines: elbow flexion/extension, joint flexion/extension of elbow and shoulder, and shoulder flexion/extension with extended forearm. These are basic exercises that are usually performed when an elbow injury occurs for example. Here we present and analyze the results of the shoulder flexion/extension with extended forearm.

In control simulation, the gain matrices K_1, K_2, K_3, K_4 are defined as

$$K_1 = 200 * diag[1,1,1,1,1,1,1]$$
$$K_2 = 200 * diag[1,1,1,1,2,1,1]$$
$$K_3 = 0.2 * diag[1,1,1,1,1,1]$$
$$K_4 = 0.2 * diag[1,1,1,1,1,1]$$

The values were fine-tuned according to the desired results obtained. For the gains of the desired inertia, stiffness and damping matrices, the following values were used:

$$K_d = 0.01 * diag[1,1,1,1,1,1]$$
$$B_d = 0.01 * diag[1,1,1,1,1,1]$$
$$M_d = 0.1 * diag[1,1,1,1,1,1]$$

Desired trajectories are periodic, proposing a frequency of $f = 0.1\ Hz$ to obtain slow movements as a regular routine. Disturbances between 0.1 to 0.5 Nm are included to evaluate the behavior of the system when external forces appear, for instance, involuntary movements of the patient due to pain. Three different tests were carried out in simulation. Here, we present results of one routine that consists in shoulder flexion/extension. The simulation results are shown in Fig.3. The resulting joint space and end-effector trajectories are shown in Fig.3a. Joint errors and torque control switching for desired forces are shown in Fig.3b. In all cases the control stabilizes at 0.9 seconds approximately without presenting overshoots despite the quick stabilization. Notice in Fig.3a there is no perceptible vibrations and errors throughout the trajectory. However, in Fig.3b (left), for constant values in the curves there are vibrations at maximum amplitudes of ± 0.015 (radians) for both joint curves and end effector curves. Moreover, the response obtained is free of noise despite the disturbances.

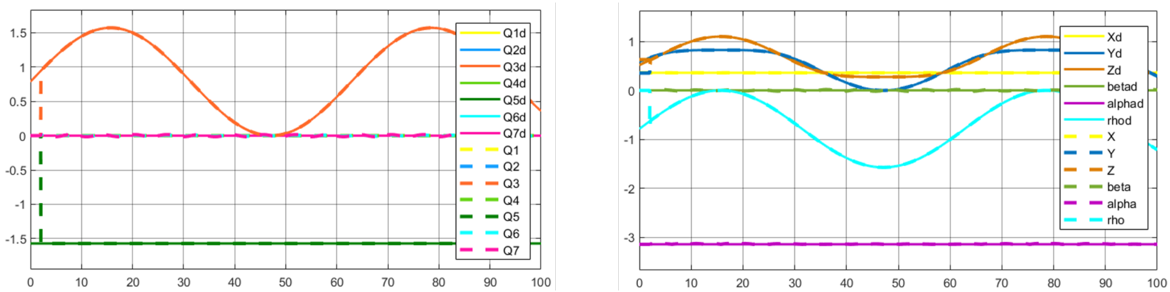
In all the tests carried out, the error is under 2%. The matrix of position-based to force-based control switched in the middle of the routine. Fig.3b (right) shows the moment when desired forces are commuted. Notice that the performance of the trajectories was preserved. For stability validation, the type of control was changed from position-based to force-based by switching the matrix S starting with desired forces of value $Fd = 1Nm$ and then, increased by an average of 6% from second $t = 50$. The trajectory does not diverge and the properties of the signal are still preserved. That means, the control is working at a different desired force without deviating from the desired trajectory of the routine. The other two routines had the same performance, the properties of the trajectory obtained were still being preserved throughout the simulation.

The obtained behavior is what is expected to be obtained in robotic rehabilitation systems due to the adaptive properties of the control to changes in position and force as in a conventional routine. These changes can also be a consequence of sudden movements that are usually produced by pain, but also occur when forces are applied progressively to the patient to increase muscle strength and on the other hand, there is the option of adaptation to any therapy under the AAN concept.

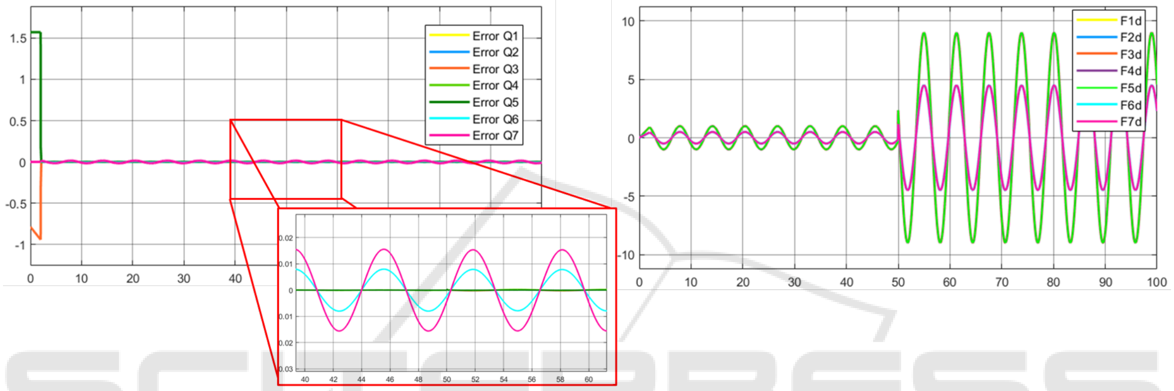
5 CONCLUSIONS

In this work, a control technique based on the combination of nonlinear adaptive control theory and hybrid impedance (position-based and force-based control) applied to a 7-DoF upper limb assistive robotic system is proposed. Hybrid impedance control is responsible for providing the levels of intervention of the assistive system to the patient for mobility recovery (position-based control) and muscular strength (force-based control).

Through nonlinear adaptive theory, we ensure the stability when changing the type of control. We performed a stability analysis using Lyapunov functions and we validated through simulations. The approach of the Lyapunov functions helps to develop the control laws, guarantee stability, and also provides adaptability to non-linearities and disturbances. Three tests were considered as part of the validation of the proposed control technique. Here we analyze the behavior of the system when performing the shoulder flexion/extension routine. The results show that the controller tracks with (almost) zero error (2%) the desired trajectories, demonstrating the effectiveness of the proposed strategy. Therefore, this control gives an insight into new opportunities for the improvement of existing or development of new strategies applied to



(a) Shoulder flexion/extension routine: joint space trajectory (left) with desired joint positions (continuous line) Q_id in radians and obtained joint position (dashed line) Q_i in radians for $i = 1..7$ and cartesian space (right) with desired cartesian positions (continuous line) X_d, Y_d, Z_d in meters and desired orientations (continuous line) $beta_d, alpha_d, rho_d$ in radians, and obtained cartesian positions (dashed line) X, Y, Z in meters and obtained orientations (dashed line) $beta, alpha, rho$ in radians.



(b) Joint errors (left) in radians and desired forces commutation (right) from position-based to force-based control

Figure 3: Shoulder flexion/extension routine: trajectory in articular and Cartesian end-effector space, joint errors and desired forces commutation from position-based to force-based control.

physical rehabilitation systems in order to guarantee safety in the interaction with patients. Future work will be focused on implementation and validation of the control in the physical rehabilitation device.

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