# Simulation and Gait-Pattern Adaptation of an Exoskeleton for Lower Limbs

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***Abstract.*** *This paper deals with modeling and simulation of an exoskeleton for lower limbs based on a commercially available orthosis. Exoskeletons are powered orthosis developed for power augmentation, for helping physically weak or injured people increase their mobility, or for rehabilitation. The main dynamic characteristics of the orthosis-patient system are considered in a MatLab-based environment simulation. Also, two gait-pattern adaptation algorithms are implemented in the proposed model, allowing patients to change the gait-pattern according to their degree of voluntary locomotor capability. In the first one, based on the inverse dynamics, the trajectory adaptation is performed by computing joint trajectories that leads to a reduction of the estimated interaction forces between the human and the robot. In the second, based on the direct dynamics, the human-robot interaction forces are translated into a desired change in the trajectory accelerations. The implementation of the gait-pattern is achieved by changing the parameters of the reference trajectories of the hip, knee, and torso joints.*

***Keywords:*** *Lower limb orthosis, Exoskeleton, Modeling and simulation, Gait-pattern adaptation*

## 1. INTRODUCTION

Amongst the diverse unfoldings studies on bipedal robots, one research area that have produced exciting results is the use of exoskeleton for power augmentation and for helping physically weak or injured people (Ferris et al. 2005, Guizzo & Goldstein 2005). Biologically, exoskeletons are external structures of insects and crustaceans that supply them support and protection. In military area, exoskeletons have being used to increase the physical capacity of soldiers, so that they can load weights that a normal personal would not obtain. Other activities of exoskeletons for power augmentation are: emergency and rescue teams can use them in rubbish areas and rugged terrain where no wheeled vehicle could move, and firemen can carry heavy equipments into burning building or injured people out of them.

Several research centers around the world are constructing exoskeletons for power augmentation. HAL-5 (Hybrid Assistive Limb), from University of Tsukuba, Japan, is the fifth generation of projected exoskeletons for power augmentation (Hayashi et al. 2005, Kim & Sankai 2005a, Kim & Sankai 2005b). It uses mioeletric signals obtained from sensors to take the necessary control action. The BLEEX (Berkeley Lower Extremity Exoskeleton), from the University of California at Berkeley, USA, uses hydraulic actuators supplied by a pump connected into a small gasoline engine (Zoss et al. 2006, Chu et al. 2005, Kazerooni 2005). More than 40 sensors together with the actuators form a local net that works as the human nervous system (Kim et al. 2004). The sensors constantly give information to the central computer that calculates the necessary action to distribute the weight in such a way that the user does not feel the exceeding weight. The exoskeleton from the Sarcos Research Corp., USA, uses force sensors in contact with the user body to supply information to the controller so that the exoskeleton can move in a harmonic way with the movement of the user. The power unit of this system is also an internal combustion engine.

In (Walsh et al. 2006), an underactuated and lightweight exoskeleton that considers the passive dynamic of walking, differently of the ones described above, is being developed. Two architectures are explored: the first one considers a spring in the hip, a variable impedance device in the knee and a spring in the ankle; the second one substitutes the spring of the hip for a no conservative actuator to examine the effect of power addition during the walking cycle.

The use of the robotics in the rehabilitation of patients with mobility dysfunctions is becoming more common, mainly due to the importance of exercises for functional rehabilitation (Ferris et al. 2005). In (Jezernik et al. 2003), a robotic orthosis, called Lokomat, is used in rehabilitation of patients with stroke or spinal cord injury individuals. The device is installed in a treadmill and the patient walks using a weight compensator. The regular training of patients is carried through imposing a fixed gait-pattern through a joint position control of the robotic orthosis. However, it is important to guarantee that the patient is effectively walking, and not only having its leg moved passively for the locomotion device.

This idea took to the development of gait-pattern adaptation algorithms (Jezernik et al. 2004, Riener et al. 2005). These algorithms make possible the patient to modify the gait-pattern as his/her degree of voluntary locomotion.

In (Riener et al. 2005), three gait-pattern adaptation algorithms are presented with experimental results from the robotic orthosis Lokomat. The first one, based on inverse dynamics, generates the adaptation of the gait-pattern estimating the human-robot interaction torques and then computing the joint trajectories in order to reduce such interaction torques. The second one relates the interaction human-robot torque with the necessary change of acceleration of the joint trajectories, using the direct dynamics. In the third algorithm, an impedance controller generates a relation between the torques of interaction and the position deviations.

This paper presents the first results of the development of an exoskeleton for lower limbs based on a commercially available orthosis. The data acquisition and control systems are being developed, considering the utilization of encoders for joint measuring, force sensors for ground reaction forces measuring, and series elastic actuators for joint actuation. Series elastic actuators are force controllable actuators which have a bandwidth similar to that of human muscle. The dynamic model of the orthosis is generated using the Matlab environment and two gait-pattern adaptation algorithms proposed in (Jezernik et al. 2004, Riener et al. 2005) are implemented, the algorithms based on inverse and direct dynamics of the orthosis-patient system.

The papers is organized as follows: Section 2. introduces the gait-pattern adaptation algorithms based on inverse and direct dynamics applied to the exoskeleton of this paper; Section 3. presents the orthosis modeling, considering the dynamic parameters computation, and the fist developments on the data acquisition and control systems of the orthosis; Section 4. presents the results of the implementation gait-pattern adaptation algorithms in the proposed exoskeleton model; and Section 5. summarizes the main contributions of the paper.

## 2. GAIT-PATTERN ADAPTATION ALGORITHMS

For the development and application of the gait-pattern adaptation algorithms proposed in (Jezernik et al. 2004, Riener et al. 2005), the orthosis is modeled using the basic robotic equation,

$$M_{ort}(q)\ddot{q} + C_{ort}(q,\dot{q}) + G_{ort}(q) = \tau_a + \tau_{pat} - \tau_{frict}, \tag{1}$$

where $q \in \Re^n$ is the generalized coordinates vector, $M \in \Re^{nxn}$ is the symmetric positive definite inertia matrix, $C \in \Re n$ is the centrifugal and Coriolis torques vector, and $G \in \Re^n$ is the gravitational torques vector. The terms $\tau \in \Re^n$ are the torques acting in orthosis: $\tau_a$ is the torque supplied by the actuators, $\tau_{pat}$ is the torque generated for the orthosis-patient interaction, and $\tau_{frict}$ is the torque generated for the friction in the joints,

The torque of interaction between the orthosis and the patient, $\tau_{pat}$, can be divided in active and passive components. The passive patient torque, $\tau_{pat,pas}$, is the torque necessary to move the patient if he/she is moving in a passive way. In case that the patient influences in the orthosis movement, it will produce the active patient torque, $\tau_{pat,act}$. Therefore, Eq. (1) can be rewrite, considering now, the orthosis-patient dynamics,

$$M_{ort+pat}\,(q)\,\ddot{q} + C_{ort+pat}\,(q,\dot{q}) + G_{ort+pat}\,(q) = \tau_a + \tau_{pat,act} - \tau_{frict}, \tag{2}$$

where $M_{ort+pat}(q)$, $C_{ort+pat}(q,\dot{q})$, and $G_{ort+pat}(q)$ correspond to the combination of the orthosis and patient dynamics. The torque to be applied by the actuators is computed in this paper by a derivative proportional controller (PD), considering the reference trajectory generated by the gait-pattern adaptation algorithms.

### 2.1 Optimization Parameters

The reference trajectories for the orthosis are parameterized in the following way,

$$\phi_i = a_i \cdot \phi_{i,n}\left(\frac{t}{c_i}\right) + d_i, \tag{3}$$

where $\phi_{i,n}$ is the nominal trajectory and $a_i$, $c_i$, and $d_i$ are the parameters used in the optimization. The parameter $a_i$ changes the trajectory amplitude, the parameter $c_i$ has influence in the step period and the parameter $d_i$ uniformly dislocates the trajectory throughout the time. Varying the $a$-$d$ parameters, different joint trajectories are generated and, consequently, different gait patterns are produced. Parameters $c_i$ are not independent, since the trajectories of all joints must have the same cycle time. After each adaptation calculation, they are substituted by their arithmetic mean.

### 2.2 Adaptation Algorithm Based on Inverse Dynamics

The active patient torques between patient and orthosis are nulls in the case the patient follows the movement assigned for the device, and different of zero in the case the patient has the intention to carry through a different movement of the programmed one. The algorithm is based on the inverse dynamics of the orthosis-patient system and works by minimizing the interaction torques, changing the desired trajectories for the orthosis.

Initially the torques the patient executes in the device is measured through a force measurement system. A variation in the reference trajectory is then computed in such a way that the variation of the torque applied for the orthosis generates a reduction in the active torque executed by the patient. The necessary calculations are made by means of nonlinear optimization of the following functional,

$$\begin{aligned} J\left(\delta q_r, F\right) &= \sum \left\| \tau_{pat,act}\left(F\right)_k - \delta\tau\left(\delta q_r\right)_{(k)} \right\|_2^2, \\ \delta q_{r-adapt} &= arg_{\delta q_r} min J\left(\delta J_r, F\right), \end{aligned} \tag{4}$$

where $F$ is the interaction forces between patient and orthosis, $\delta q_r$ is the reference trajectory change.

In Eq. (4), the torque variation, $\delta\tau$, produced by a change in the reference trajectories, results in a reduction of the patient active torque, $\tau_{pat,act}$. $\delta\tau$ is calculated from the orthosis-patient dynamics,

$$\begin{aligned} \delta\tau\left(\delta q_r\right) =& M_{ort+pat}\left(q_r + \delta q_r\right)\left(\ddot{q}_r + \ddot{\delta} q_r\right) + C_{ort+pat}\left(q_r + \delta q_r, \dot{q}_r + \dot{\delta} q_r\right) \\ &+ G_{ort+pat}\left(q_r + \delta q_r\right) - M_{ort+pat}\left(q_r\right)\ddot{q}_r - C_{ort+pat}\left(q_r, \dot{q}_r\right) - G_{ort+pat}\left(q_r\right), \end{aligned} \tag{5}$$

where $q_r$, $\dot{q}_r$, and $\ddot{q}_r$ are the reference trajectory and its derivatives, respectively.

As gait-pattern adaptation must occur in real time, a fast and simple minimization algorithm must be used. The method of the steepest descent, that considers the first derivative, converges slowly in the neighborhood of the minimum, but it is powerful for distant configurations of a minimum. Hence, it was selected to calculate the parameters that minimize the functional $J$. The parameters are updated according to the following equation,

$$p_{n+1} = p_n - \eta_{p,n} \frac{\delta J}{\delta p}|_{p_n}, \quad p \in \{a_1, c_1, d_1, ..., d_6\}, \tag{6}$$

where $\eta_p$ is the step size and $n$ is the number of necessary iterations. The step size is taken as a descent geometric sequence,

$$\eta_{p,n+1} = q.\eta_{p,n}, \quad q < 1. \tag{7}$$

The $n = K$ iterations are carried through after the choice of initial parameters $(a_{i,0} = 1, c_{i,0} = 1 and d_{i,0} = 0)$ and of the initial step $\eta_{p,0}$.

### 2.3 Adaptation Algorithm Based on Direct Dynamics

To reduce the dynamic model dependence in the adaptation algorithm of Section 2.2, the proposed algorithm is changed considering the direct dynamics of the system orthosis-patient. Consider the acceleration of the generalized coordinates,

$$\ddot{q} = M^{-1}_{ort+pat}\left(q\right)\left\{\tau_a - \tau_{frict} - C_{ort+pat}\left(q, \dot{q}\right)\right\} + M^{-1}_{ort+pat}\left(q\right)\tau_{pat,act}. \tag{8}$$

Inspecting Eq.(7), it can be observed that the last term represents the variation in the acceleration of the reference trajectory imposed for the patient. This hypothesis forms the base of the adaptation algorithm based on direct dynamics. The idea is to estimate the term $\tau_{pat,act}$ via force measurement (the same used in Section 2.2) and calculate the variation in acceleration. Adding the measured variation to the nominal acceleration, it is gotten the patient desired acceleration,

$$\ddot{q} = -M^{-1}_{ort+pat}\left(q\right)\delta\tau_{pat,act} \Rightarrow \ddot{q}_{des,pat} = \ddot{q}_{nom} + w\ddot{\delta}q, \tag{9}$$

where $w$ is an adaptation constant. The adaptation of parameters $a$ and $c$ is carried through via minimization of the following functional,

$$J\left(\delta q_r, F\right) = \sum \left\|\ddot{q}_{des,pat}\left(F\right) - \ddot{q}_{adap}\left(\delta q_r\right)\right\|_2^2. \tag{10}$$

As no information regarding $d$ is found at the acceleration vector, due to double differentiation of Eq. (3), this parameter is adapted by minimization of the functional presented in Eq. (4). The minimization was carried through as in Section 2.2, using the method of the steepest descent. An important advantage of the direct dynamics in relation to the indirect one is the less dependence of the model; only the knowledge of inertia matrix $M_{ort+pat}\left(q\right)$ is necessary, in contrast of the indirect method that needs the knowledge of the matrices $M_{ort+pat}\left(q\right)$, $C_{ort+pat}\left(q, \dot{q}\right)$, and $G_{ort+pat}\left(q\right)$.

The gait-pattern adaptation algorithms described above can be represented by the block diagram of Fig. 1.

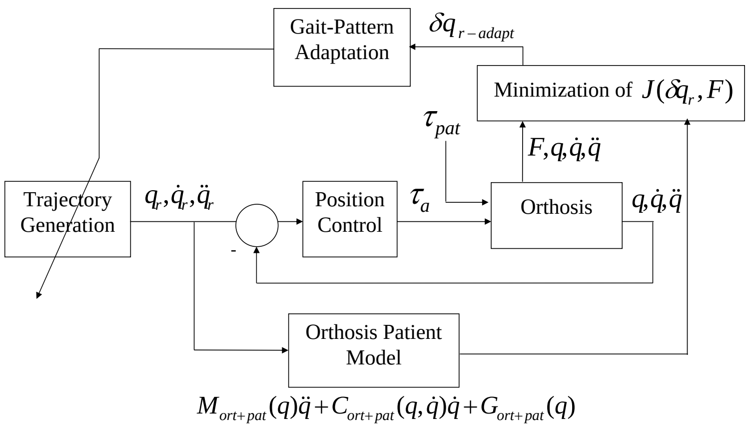


Figure 1. Gait-pattern adaptation algorithms block diagram.

## 3. ORTHOSIS MODELING

The orthosis that will be used trough the main project of construction of an exoskeleton for lower limbs, Fig. 2, corresponds to one Reciprocating Gait Orthosis LSU (Lousiana State University), developed in the Ontario Crippled Children's Centre, in Toronto, Ontário, at the beginning of 1970. Indicated for carrying paraplegic patients with spinal cord injuries, the orthosis is composed of two KAFOs (Knee-Ankle-Foot Orthosis), confectioned in thermoplastic joined to a metallic pelvic band by metallic bars. Two reciprocating handles connect the pelvic joints, allowing alternated movements of flexion and extension of the hip (Carvalho 2006). The reciprocation mechanism will not be considered in the final exoskeleton, since it was verified that the real mechanism does not present good performance and its functionality can be reproduced by means of opposed action of the hip joint actuators. However, information obtained from a patient using the orthosis with the reciprocation mechanism will be considered in the exoskeleton's control system.

Figure 2. Current orthosis status and Solid Egde model.

Currently, the project of construction of the exoskeleton is in the phase of encoders installation for the accomplishment of measuring the joint angles, development of a platform for measurement of the ground reaction forces, using force sensors FRS (Force Resistor Sensor), and acquisition of a gyroscope for body orientation measuring. Therefore, it will be possible to use the exoskeleton for acquisition of the data of a patient using it, with the reciprocation mechanism and without it. These data will be used as normal trajectory for the gait-pattern adaptation algorithms. For joint actuation, series elastic actuators, developed in (Pratt & Williamson 1995, Robinson et al. 1999), are being reproduced. Series elastic actuators are force controllable actuators which have a bandwidth similar to that of human muscle and can be used as a variable impedance device, a natural characteristic observed in the gait-pattern. In (Pratt et al. 2004), series elastic actuators are used in the development of a device for power augmentation of the knee joint.

To obtain the dynamic parameters of the orthosis and to help on the further project developments, a Solid Edge representation of it was built, considering some simplifications due to the orthosis' complex forms, Fig. 2. The values of the orthosis parameters are shown in Tab. 1. It is also presented the parameters of the patient considered in the simulation, obtained from (Winter 1990), considering a 85 kg, 1.74 m individual.

Table 1. Orthosis and Patient Dynamic Parameters.

| **Orthosis** | | **Patient** | |
|---|---|---|---|
| $M_{total,ort}$ | 4.8 | $M_{total,pat}$ | 85 |
| $L_{total,ort}$ | 1.0 | $L_{total,pat}$ | 1.74 |
| **Limb Mass (kg)** | | | |
| $M_{tigh,ort}$ 0.95 | | $M_{tigh,pat}$ | 8.5 |
| $M_{leg+foot,ort}$ | 0.72 | $M_{leg+foot,pat}$ | 5.2 |
| $M_{torso,ort}$ | 1.49 | $M_{torso,pat}$ | 57.6 |
| **Limb Length (m) - z direction** | | | |
| $L_{tigh,ort}$ | 0.39 | $L_{tigh,pat}$ | 0.39 |
| $L_{leg+foot,ort}$ | 0.49 | $L_{leg+foot,pat}$ | 0.49 |
| $L_{torso,ort}$ | 0.12 | $L_{torso,pat}$ | 0.87 |
| **Limb Center of Mass (m) - z direction** | | | |
| $CM_{tigh,ort}$ | 0.18 | $CM_{tigh,pat}$ | 0.17 |
| $CM_{leg+foot,ort}$ | 0.17 | $CM_{leg+foot,pat}$ | 0.30 |
| $CM_{torso,ort}$ | 0.11 | $CM_{torso,pat}$ | 0.33 |
| **Limb Inertia Momentum ($kgm^2$) - sagital plane** | | | |
| $I_{tigh,ort}$ | 0.03 | $I_{tigh,pat}$ | 0.14 |
| $I_{leg+foot,ort}$ | 0.02 | $I_{leg+foot,pat}$ | 0.22 |
| $I_{torso,ort}$ | 0.06 | $I_{torso,pat}$ | 10.73 |

An analytical model of the orthosis, considering the patient interaction and ground reaction forces, is developed using the Symbolic Toolbox of the Matlab. The generalized coordinates used in the orthosis-patient model can be seen in Fig. 3. The absolute angles with relation to the vertical line were considered: trunk ($q_{torso}$), right thigh $\left(q_{fem1}\right)$, right leg ($q_{tib1}$), left thigh $\left(q_{fem2}\right)$, left leg ($q_{tib2}$) and all the body, considering the lateral motion ($q_s$).

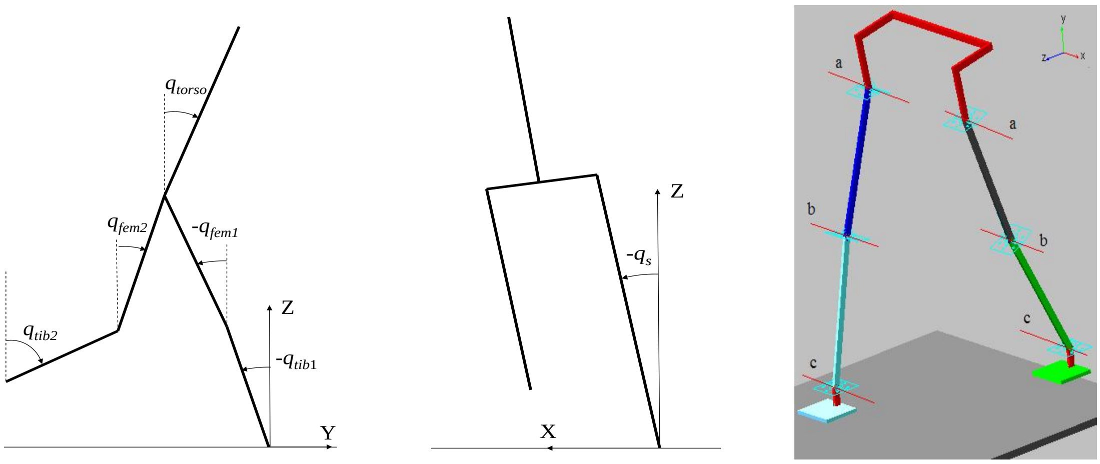


Figure 3. Generalized coordinates of the orthosis-patient model and ADAMS/MSC model.

It is also been developed, using the ADAMS/MSC simulation environment, a complete model of the system, considering the patient and the orthosis, and the interaction forces between them and the ground. Figure 3 shows the first model constructed, considering only the orthosis. Future works consist in the development of a representation composed of a human body wearing the exoskeleton. The results of the gait-pattern adaptation algorithms implementation, described in the next section, were obtained using the Matlab environment.

## 4. RESULTS

In this section, the two gait-pattern adaptation algorithms presented in Section 2. is implemented in the model of the orthosis of Figure 2. Since no clinical gait analysis have been performed yet for a patient using this device, the nominal trajectory used in Eq. (3) is obtained from the results presented in (Kirtley n.d.). In that work, joint positions and velocities for 20 individuals (10 young, from 15 to 25 years, and 10 adults, from 45 to 55 years) were measured during walking with average speed of 1.07 m/s. Joint torque and power consumption are also available.

As only simulation is performed in this work, the interaction torques between orthosis and patient (active patient torque) were artificially estimate from a given trajectory. In this work, this value is computed through the comparison between the desired trajectory for the patient, $q^d_{pat}$ (defined previously by the simulator user) and the real trajectory, $q$. It is assumed that the interaction torque results of a spring type virtual coupling between the patient desired position and the real position,

$$\tau_{pat,act} = K\left(q^d_{pat} - q\right). \tag{11}$$

The spring stiffness is adjusted in order to take realistic magnitudes of the active patient torque. The values of the parameters, $a$, $c$ and $d$ in the desired trajectory of the patient are shown in Tab. 2. The values of $d$ for all the angles correspond to an addition of 4° degrees in the trajectory. The values of $c$ correspond to a reduction of 5% in the total step time.

Table 2. Patient Desired Trajectory Parameters.

| | $a$ | $c$ | $d$ |
|---|---|---|---|
| $q_{tib1}$ | 1.05 | 1.05 | 0.069 |
| $q_{fem1}$ | 1.05 | 1.05 | 0.069 |
| $q_{torso}$ | 1.05 | 1.05 | 0.069 |
| $q_{fem2}$ | 1.05 | 1.05 | 0.069 |
| $q_{tib2}$ | 1.05 | 1.05 | 0.069 |
| $q_s$ | 0.95 | 1.05 | 0.069 |

The adaptation of the parameters, $a$, $c$ and $d$, resultants of the optimization of the functional $J$, is carried through independently for each joint (left and right knee, left and right hip, trunk, and lateral movement). To keep symmetrical walking, only the joints of the swing leg is considered in the optimization process, the parameters of the stance leg is assumed equal to the ones of the swing leg. Each sequel of optimization is conducted at the end of the step (off-line, this procedure is possible in this type of adaptation), considering three equally spaced points throughout the step time.

The values of the parameters are then updated as nominal trajectory and a new simulation of the orthosis-patient model is performed. As observed in Section 2, the values of the parameter $c$ must be adjusted, through an arithmetic mean, so that all joint trajectories have the same cycle time. The value of $w$ used in the algorithm based on the direct dynamics is 0.5. Table 3 shows the final adapted parameter values after three steps, that is, after three sequences of optimization of the parameters. Figures 4 and 5 present the nominal, patient desired, adapted and real trajectories for each joint, referring four steps, initiating with the right leg in stance, for the method based on the inverse and direct dynamics, respectively.

Table 3. Adapted Trajectory Parameters.

| | **Inverse Dynamics** | | | **Direct Dynamics** | | |
|---|---|---|---|---|---|---|
| | $a$ | $c$ | $d$ | $a$ | $c$ | $d$ |
| $q_{tib1}$ | 1.0483 | 1.0493 | 0.0695 | 1.0484 | 1.0581 | 0.0711 |
| $q_{fem1}$ | 1.0502 | 1.0493 | 0.07 | 1.0487 | 1.0581 | 0.0659 |
| $q_{torso}$ | 1.0495 | 1.0493 | 0.0683 | 1.0499 | 1.0581 | 0.0689 |
| $q_{fem2}$ | 1.0502 | 1.0493 | 0.07 | 1.0487 | 1.0581 | 0.0659 |
| $q_{tib2}$ | 1.0483 | 1.0493 | 0.0695 | 1.0484 | 1.0581 | 0.0711 |
| $q_s$ | 0.9479 | 1.05 | 0.0693 | 0.9538 | 1.0581 | 0.0690 |

Note that as the second step has less duration in relation to the first one, because the desired trajectory for the patient for joint $q_s$ (addition of 4 degrees in parameter $d$) does an inclination of all body for the right side, that makes the right leg, in swing phase, touches the ground before the desired time. Comparing the two algorithms from Fig. 4 to 5 and Tab. 3, it can be observed that both algorithms obtained satisfactory results with relation to the adaptation of the parameters used in the patient desired trajectory. Also, note that the algorithms have good results just at the beginning of the second walking cycle (after two steps). Thus, the necessary time for the adaptation of the trajectory is small, showing the functionality of the algorithms.

## 5. CONCLUSIONS

This paper presents the first results of the development of an exoskeleton for lower limbs based on a reciprocating gait orthosis. A complete model and a gait environment simulation of the orthosis are obtained using the symbolic tools of the Matlab language. It is also presented the implementation of two gait-pattern adaptation algorithms in the proposed

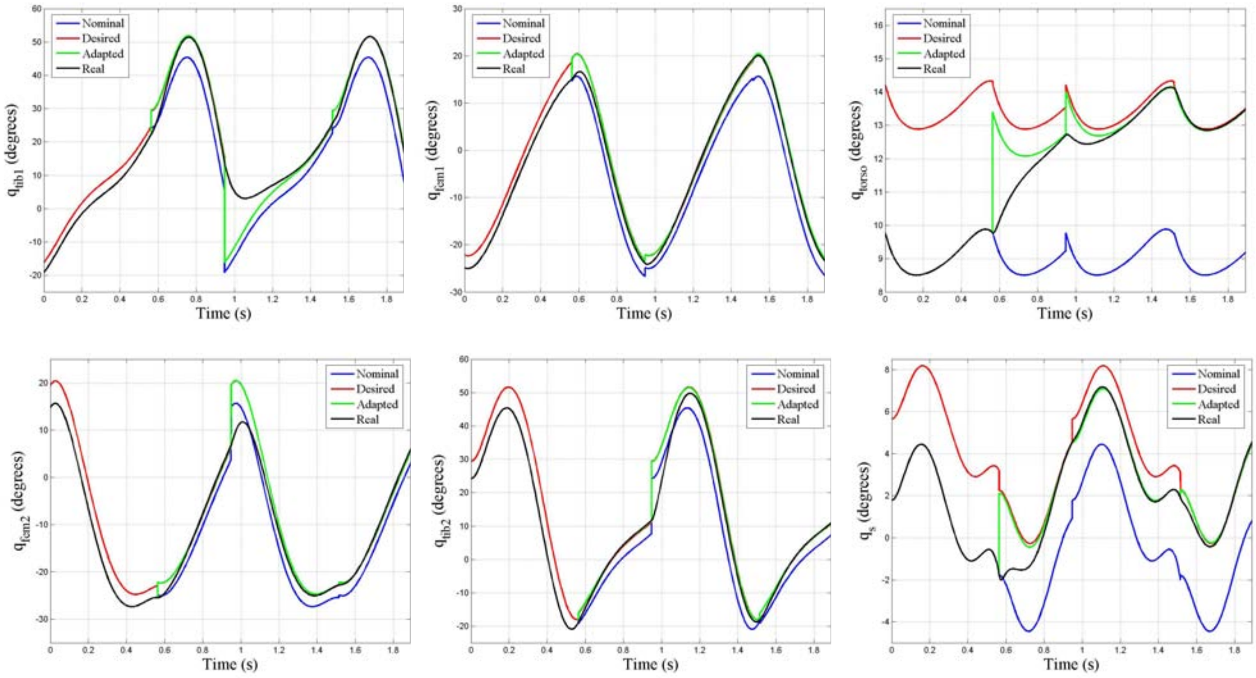


Figure 4. Nominal, patient desired, adapted and real trajectories for each joint, method based on the inverse dynamics.

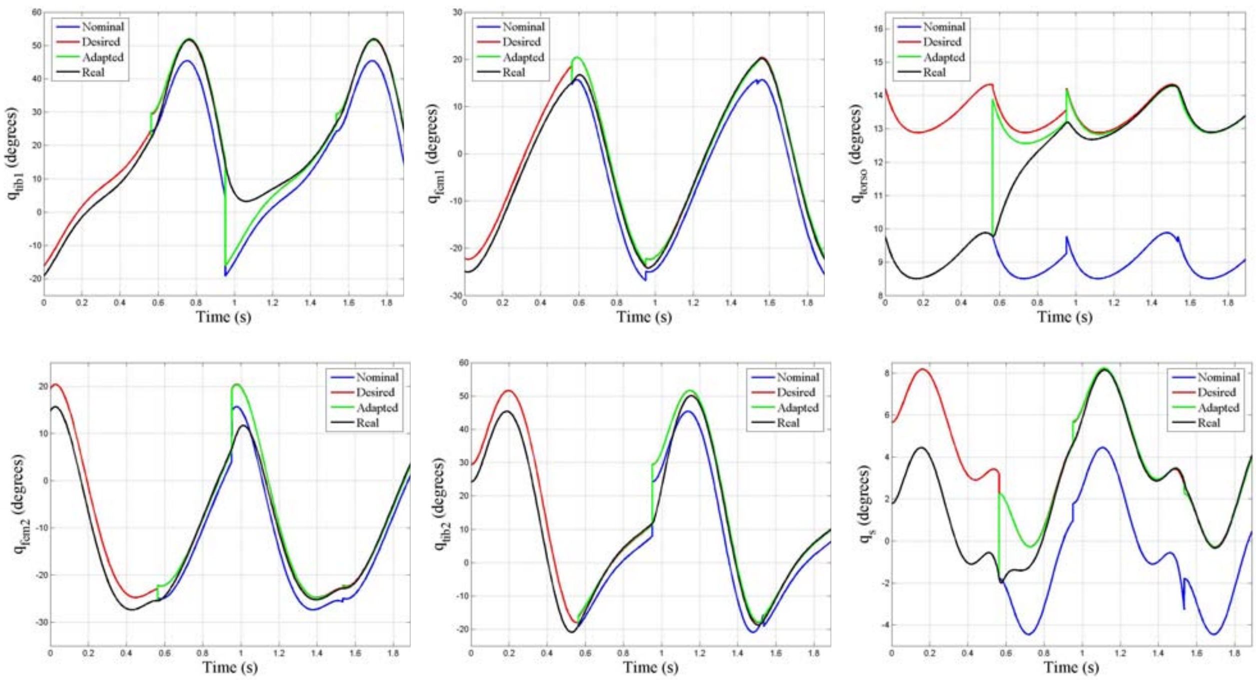


Figure 5. Nominal, patient desired, adapted and real trajectories for each joint, method based on the direct dynamics.

orthosis-patient system, allowing the patient to modify the gait-pattern as his/her degree of voluntary locomotion. The results obtained show that the proposed adaptation algorithms can be applied in the actual exoskeleton being constructed and obtain, considering the experimental implementation problems, satisfactory performances.

## 6. ACKNOWLEDGEMENTS

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