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A motor learning oriented, compliant and mobile Gait Orthosis

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Abstract. People affected by Cerebral Palsy suffer from physical disabilities due to irreversible neural impairment since the very beginning of their life. Difficulties in motor control and coordination often relegate these patients to the use of a wheelchair and to the unavoidable upcoming of disuse syndromes. As pointed out in recent literature Damiano [7] physical exercise, especially in young ages, can have a deep impact on the patient health and quality of life. For training purposes is very important to keep an upright position, although in some severe cases this is not trivial. Many commercial mobile orthoses are designed to facilitate the standing, but not all the patients are able to deploy them. ARGO, the Active Reciprocated Gait Orthosis we developed, is a device that overcomes some of the limitations of these devices. It is an active device that is realized starting from a commercial reciprocated Gait Orthosis applying sensors and actuators to it. With ARGO we aim to develop a device for helping limbs in a non-coercive way accordingly to user's intention. In this way patients can drive the orthosis by themselves, deploying augmented biofeedback over movements. In fact Cerebral Palsy patients usually have weak biofeedback mechanisms and consequently are hardly inclined to learn movements. To achieve this behavior ARGO deploys a torque planning algorithm and a force control system. Data collected from a single case of study shows benefits of the orthosis. We will show that our test patient reaches complete autonomous walking after few hour of training with prototype.

Keywords: Rehabilitation robotics, active orthoses, cerebral palsy, pneumatic artificial muscles, force control

1. Introduction

Cerebral palsy (CP) is an umbrella term encompassing a group of non-progressive, non-contagious motor conditions that cause physical disabilities in human development, chiefly in body movement and positioning. Typical symptoms of CP are abnormal muscle tone, abnormal stretch reaction, anomalous motor development and difficulties on coordination. CP is classified into four major areas to describe different movement impairments (spastic, ataxic, atheoid/dyskinetic and hypotonic). They are all characterized by abnormal muscle tone, reflexes, difficult motor development and lack of coordination. The outcome is variable from each patient. In some case it drives to a constant augment of the muscle tone, without the ability to control it, while in other patients the muscles are so weak that they are unable to support patient's weight. CP is not a progressive disorder, and therefore the syndrome does not get worse neither better with time. However, a patient experiencing motor difficulty tends to assume unnatural postures and adopt a sedentary life style, which can cause serious health problems. Moreover, while wheel chairs are normally used to help the movement of people with gait disorder, they cause a series of complications known as disuse

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syndrome. Although there is no cure for CP, physical therapy helps the body to maintain muscle tone, bone structure, and prevents dislocation of the joints.

An orthosis is an assistive device used to support part of a patient body and to help it work better. Focusing on lower limb orthoses, many devices have been developed so far for gait assistance/training in CP affected people. While passive orthoses apply mechanical restrictions to the joints to help user gait, active gait orthoses integrate actuators to support patient gait. By active control of joint torques, these mechanisms are expected to help patients who are not able to walk with the aid of passive orthoses. Active assistive orthoses tackles the disuse syndrome by helping the user to keep an upright position, walk, and improve patient gait efficiency.

This paper describes the study and development of an active Gait Orthosis with compliant characteristics and intelligent control of actuators. In fact the prototype is based on pneumatic artificial muscles (PAM) which are inherently safe and compliant. On the other hand these actuators are not easy to control because of non-linear dynamics [4, 13, 20].

In section 2 we present an overview of the state of the art concerning this work: cerebral palsy physical treatments and active orthoses. Section 3 introduces our prototype, dividing the description into three conceptual parts: hardware, control and planning. Section 4 focuses on some details of control. As a validation, in section 5 we present data collected during treatment with a test patient. Patient conditions before and after treatment are discussed considering data from EMG and metabolic cost analysis. Finally section 6 presents conclusions and our plans for future work.

2. State of the art

2.1. Cerebral palsy

Cerebral palsy has an incidence of birth between 0.15 and 0.25% and it can exhibit in various ways leading to very different case histories and inability levels depending on localization an extension of lesions in central nervous system. Despite treatments can consequently be very different, it is proven that precocity of rehabilitation has a fundamental role in prevention of secondary deformity and anomalous motor development [2]. Also recent studies pay great attention on physical therapy applied to young CP patients, focusing on movement based strategy and physical training. As example Dodd [8] proves that resistance training is able to increment muscle force in CP patient. Analogous very encouraging results are about protocols of training and physical therapy using a treadmill [7]. However these kinds of treatments are quite expensive because they need the presence of one or more physiotherapists controlling the patient safety and avoiding pathological compensations. Active or passive orthotic systems try to help these treatments relieving physiotherapist of part of work.

2.2. Active orthoses

Commercial active orthoses are nowadays quite common. They can be classified in external and peripheral actuated. The first ones are more sophisticated and versatile, they use actuators for every moving joint. On the other hand peripheral actuation attempt to carry only the extremities without considering entire limb kinematics. Examples of external actuated orthosis are Lokomat® and ReWalk™ whereas Gangtrainer™ and Innowalk® are peripheral actuated. Unfortunately most of these commercial orthoses are not mobile because their mass and power needs. However some of them overcome this limitation by deploying a virtual reality environment to augment patient motivation.

The field of active orthosis on the research side is very wide. It starts on the previous half century with purely mechanical orthoses, reaching the first computer based orthosis around the 70s years. For a more detailed historical background and state of the art of active orthoses see Dollar and Herr [9].

Some example of mobile and external actuated orthoses are for example the AAFO, Active Ankle Foot Orthosis, which assists the drop-foot gait by modulating the impedance of the orthotic joint throughout the walking cycle [1]. AAFO misses full leg support, which is indeed provided by the Wearable Walking Helper (WWH) or the commercial ReWalk™, two active hip-knee orthosis that can help the user in common daily movements like walking, doing steps, standing and sitting (see [11, 12]). Both of them require the user to keep balance by himself, or with external support (crutches). If this is not possible, either because of the weakness of the patient or his/her difficulty in movement control, a more stable orthosis is needed. Many active gait orthoses demand the user to hang in the front or the backside in order to hold trunk in an erect posture. SUBAR (Sogang university biomedical assist robot, Kyoungchul [16]), makes use of a self-moving appendix located in front of the wearer, to sustain the patient and carry heavy peripheral devices. The need to grab a support has the inconvenience of inducing forward or backward tilting of the upper half body, making it difficult to keep right posture for walking.

3. The prototype

ARGO, the Actuated Reciprocated Gait Orthosis (Fig. 1), is the mobile active orthosis developed the robotics laboratory ALTAIR of the University of Verona. His structure is based on a commercial passive Reciprocated Gait Orthosis (RGO), the NF-Walker® (NFW), which has been modified to accommodate sensors and actuators. The idea to start from a commercial orthosis comes from the fact that it has already been optimized in the mechanical coupling between human body and reciprocation, and has already been widely tested by physicians and patients. The choice of using this specific orthosis comes from the mechanical characteristics of the device and from the type of interaction with the human wearer. The patient hangs from the orthosis, without the danger of falling, and his/her weight is partially sustained by the orthosis and partially by the patient own legs. The orthosis is mobile and provide reciprocation so when a leg goes forward the other is compliantly constrained do move backward. The wearer needs just to lift a leg, both lower limbs will move properly to perform a step, causing the orthosis to move forward.

Fig. 1. The ARGO prototype.

In literature the same orthosis is deployed by Kobayashi et al. [15]. Their prototype was intended for people with no force at all on their lower limbs. In fact they formalize a normal gait and replicate it on the system, applying a position-based control, purely targeting a fixed gait without any patient specific behavior. ARGO instead is intended for patients who can still apply a certain amount of force with their legs, at least enough to slightly rise up a feet. Thus, we want to motivate the patient to use his/her muscles by amplifying or facilitating movements, rather than binding the trajectory to a fixed model. In fact, although it is possible to find some common pattern on CP patient walk (crouch gait, scissor walking or toe walking), the actual walk varies a lot from patient to patient. Each walking strategy comes from the adaptation of the patient to his own body's pathology, which is highly specific. Thus, forcing the user to adopt a normal walk would be not only useless, but possibly even dangerous for the patient. These considerations drove us to develop a non-coercive device, able to detect patient condition and intention and generate a proper torque contribution to the hip joints. This is made possible by deploying a force control loop instead of a position control loop.

The choice of hip joints comes from the study of human walking in research on active orthosis and exoskeletons. Dollar and Herr [9] report that, for normal speed walking, aid can be considered as additive power to the hip (actuation), dissipating power at the knee (brake) and storing energy at the ankle (elastic).

Conceptually our prototype can be represented as three layers: the hardware, the control and the planning layer. The hardware layer includes the mechanical structure, actuators and electronics. The control layer consists of an force control system that employs a neural network to overcome actuators non-linearity. Finally the third layer is the only patient specific layer where force profiles can be computed according to sensors input and a finite state machine able to recognize the walking phases. In particular this first prototype is developed targeting a young patient that suffers of cerebral palsy.

3.1. Hardware layer

ARGO prototype consists of:

- The NF-Walker® orthotic system
- A pair of Festo DMSP-20 muscle
- A pair of Festo MPPE servo valve

- Four high precision potentiometer
- A pair of GICAM AF22 load cell
- A micro-controller board
- Six force sensing resistors (FSR)
- The electronic circuits for conditioning FSR and drive Festo valves
- The pneumatic circuit
- Mechanical supports for actuators, pneumatics and electronics

In the prototype Festo PAM are used as actuators, mounted in a non-antagonist configuration, exploiting the NFW mechanical reciprocation. They are connected to the NFW hip joint by a lever while a mechanical support at the backside is used to carry two pneumatic valves and all the electronic components. The muscles are attached to the device so that a contraction of the PAM causes hip flexion. Four potentiometers are mechanically fixed to hip and knee joints, two load cells are mounted on the muscle terminal nodes and three force sensing resistors are embedded in each user shoe. All the sensors are connected to a micro-controller board. The system uses a 16 MHz processor basing on single task architecture to get a reliable hard real-time computation. We came to a 60 Hz task frequency. Analog outputs are obtained by using pulse width modulation with a series analog second order RC filter at 30 Hz. Other conditioning electronics is simply based on operational amplifiers and voltage dividers. High-level features (like as data storage) are provided by a non real-time Linux embedded system based on a 400 Hz Pentium IV. Communication with the micro controller is serial. The pneumatic circuit is composed by the pneumatic source (we use a compressor, but high pressure air cylinder can be more suitable for mobility). A 1/8′ tube connects it to both the servo valves inputs using a "T" junction. The valve output is finally connected to the artificial muscles with very short tubes, so we can neglect pneumatic dynamics, which is non-linear, very hard to model and grows with tube length.

3.2. Control layer

Force control system is very important in motion assistive devices because the slow actuator response can obstruct user's movement. Control becomes critical if highly non-linear actuators such as pneumatic muscle are used. It is not possible to obtain a fast and stable controller that meets these requirements with classical techniques. In particular high gain PID lead to quick response, but cause oscillations in the system while low gain PID have smooth but too slow response. To overcome these limitations we have previously presented a method that improves performances with respect to classical PID [3].

In non-linear system control a common strategy is to build a model-based controller which takes explicitly into account non linearity. A simple way to do this is to have a feed-forward action that, using the inverse model, pre-calculates the correct pressure to obtain the desired force. Due to the non-triviality of deriving a model for the Festo muscles, our approach is to use a small neural network to predict muscle pressure. This method leads to strong improvement of control performance: low oscillation and a fast response. Because of the importance of this layer we dedicate a further section 4 to a more detailed description of the proposed control system.

3.3. Planning layer

Walking Assistive Algorithm has the responsibility of generating torque profiles and over-imposing them to the force control system. To realize a non-coercive action, we use an algorithm for walking phase recognition, based on a finite state machine (FSM). Then accordingly to FSM state, hip angle and ground reaction forces reference torque is computed. From here on we will refer to hip flexion angles considering the angle between the leg and the perpendicular to the ground. A flexion will cause an increase of the angle, while an extension will reduce it. As zero reference angles we will consider the perpendicular to the ground.

Figure 2 shows the FSM states and transitions. C1, C2 and C3 are the transitions which are symmetrical for right and left leg. Transition C1 and C3 are quite easy to detect basing only on ground reaction forces, measured via FSR sensors. On the other hand C2 is more complex. It should define when to invert the force aid to the user, from the support to the hip flexion to letting the user lower down the leg. Through patient walk observation we reach a set of assumption that can helps in detection of C2:

- In ordinary walk, the patient does not need to stand on one leg for more than a short time, he puts the other leg down to continue the gait or stop, so after three seconds of RU C2 is automatically active;

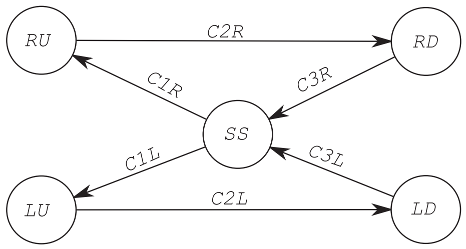

Fig. 2. Finite state machine for the walking phase recognition. Considering the NF-Walker embedded reciprocation, user's legs configuration is always compliantly constrained to the following states: Steady State (SS), Right Up (RU), Right Down (RD), Left Up (LU), Left Down (LD). C1, C2 and C3 define the transition condition, which are symmetrical for the right and left states. They are based on sensors (hip angle and ground reaction force) and leg dynamics.

- Observing the patient hip flexion capabilities we can determine a maximum angle; after reaching that angle the flexion can be considered concluded and C2 is activated;
- If the force sensor detects a force that is greater, by a threshold, than the approximate child passive leg dynamics, then it means that the patient is trying to turn down the leg and C2 is activated;
- If the hip angular velocity becomes negative, the user is lowering the leg so C2 transition is activated;

All these conditions are translated into code for switching between RU and RD and symmetrically for LU and LD. To implement the other transitions we basically use thresholds on FSR sensors embedded in user's soles. We use three sensors for each shoe for have a measure of entire or partial foot support. We provide on line normalization of these signals combining the six sensor output in a unique variable called η which is positive if the weight is distributed mainly on right foot and negative otherwise.

To support the patient we use two force generation mechanisms. The first one is the gravity compensation modelled by the following equation:

$$G(\theta_{hip}, \theta_{knee}) = K_G \left[m_p g b_p \sin\theta_{hip} + m_d g l_p \sin\theta_{hip} + m_d g b_d \sin(\theta_{hip} + \theta_{knee}) \right] \quad (1)$$

where m is the segment mass, l is the segment lenght and b is the distance between segment center of mass and upper rotation axis. The subscript p is referred to the proximal segment of the leg, while d to the distal one including the foot. Note that gravity compensation is useful only in the lifting up phase whereas during the lowering down ti is counterproductive so it must be turned off. The other force we use to help the patient is supplied according to the equilibrium of the user. For CP patient is very important to accompany and amplify natural movements. So we propose a force contribution that is computed by the following equation:

$$A(\eta) = K_A(1 - abs(\eta)) \quad (2)$$

The general effect of this contribution is to help the patient to slightly lift up a leg in the moment he is moving his weight to the opposite limb. K_G and K_A are tunable constants which depend on how much we want to influence patient movements. These two contributions can be turned on or off depending on the current state of the FSM in Fig. 2. The proposed strategy is reported in algorithm 1, where ε is a very low torque that is needed only to keep the orthosis joints in contact with the child's leg, otherwise bending of the muscle can occur. The symbol q is referred to Lagrangian

Algorithm 1 Torque reference generation algorithm

$\tau^R_{ref} = \varepsilon$
$\tau^L_{ref} = \varepsilon$
if SS **then**
  if $\eta >= 0$ **then**
    $\tau^R_{ref} = \varepsilon + A(\eta)$
  else
    $\tau^L_{ref} = \varepsilon + A(\eta)$
  end if
end if
if RU **then**
  $\tau^R_{ref} = G(q_R) - G(q_L) + A(\eta)$
end if
if LU **then**
  $\tau^L_{ref} = G(q_L) - G(q_R) + A(\eta)$
end if

coordinates ($\theta_{hip}, \theta_{knee}$) and subscripts R and L are referred to right and left leg respectively.

4. Control system details

In this section we briefly present a model-based control system to drive Festo pneumatic muscles. This is not a trivial issue because classic control is not working well with this kind of non-linear actuator. Also in practice determining PID gains by trial and error process was unsuccessful: we cannot obtain a controller fast enough and oscillation free. Figures 3–6 show some control results using PID running on a 60 Hz task. Anti wind-up schema, limit to derivative action and force noise filtering (25 Hz cut-off frequency) were implemented. Data is taken from experiments on a test bed where we use elastic rubber bands as muscle antagonist. Note that the noise amplitude depends on the entire controlled system dynamics; in this case elastic bands are very much faster than a human leg and cannot similarly filter the noise.

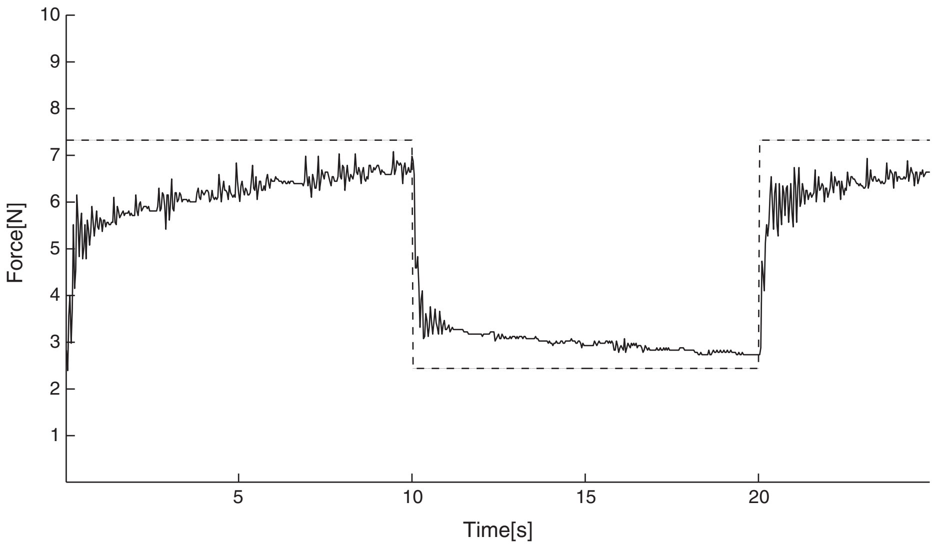

Fig. 3. Low gain PID on square wave trajectory.

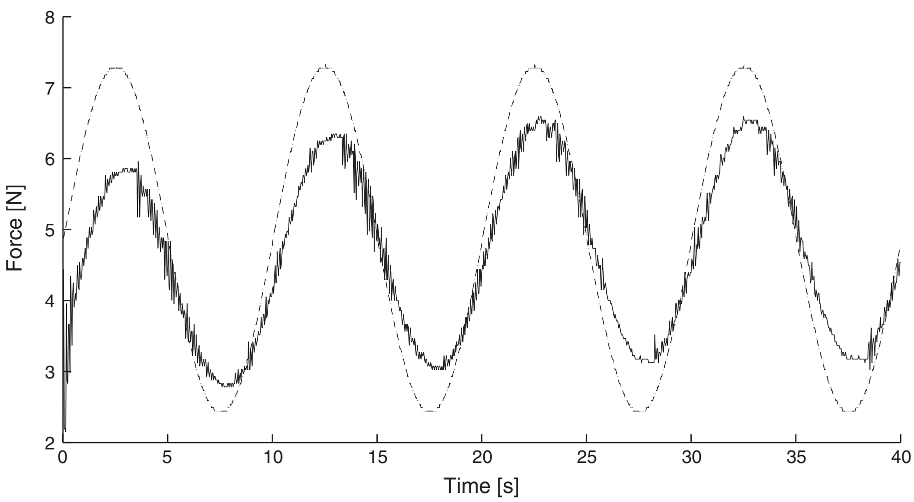

Fig. 4. Low gain PID on 0.1 Hz sine trajectory.

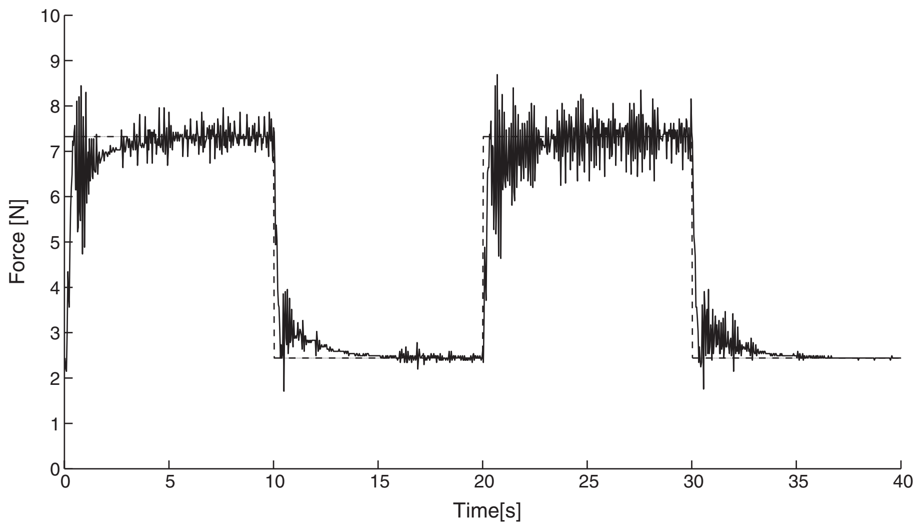

Fig. 5. High gain PID on square wave trajectory.

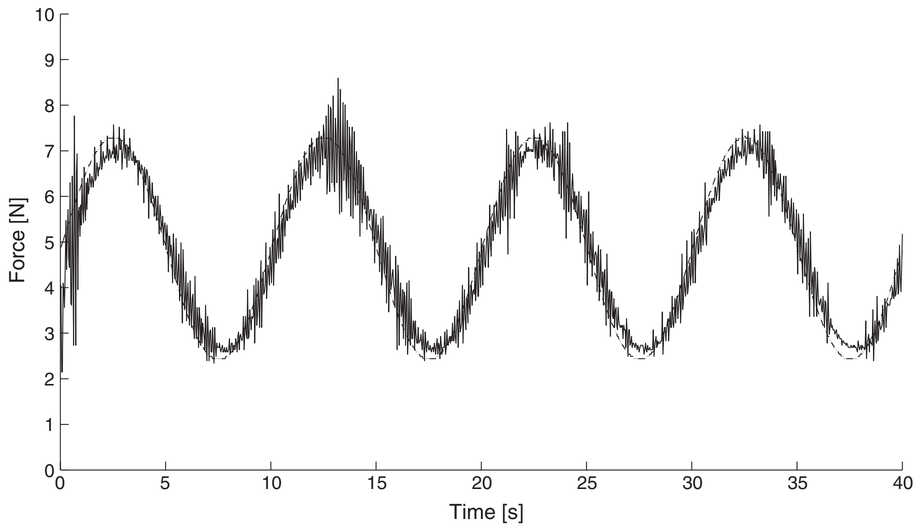

Fig. 6. High gain PID on 0.1 Hz sine trajectory.

As a single PID is not able to proper control the system we propose a model-based feed-forward (FF) action that take into account muscle non linearity. Basically FF action predicts an approximate value for the target pressure in the muscle, and then PID control just needs to refine it so that fast control dynamics is no longer necessary. The first thing we have to point out about the model based part is that classical McKibben models by Chou and Hannaford [5], Chou and Hannaford [6] and Surentu et al. [19] are not suitable for Festo muscle, as we already shown in [3]. This is probably due to the different mechanical structure of the actuators. In fact in classical McKibben's the braid and the fibres are two different sliding layer while in Festo's there is a unique layer of mixed rubber and fibres. This drove us to the necessity of using a non-physical based model. Some attempt to find suitable polynomial models can be found in Knestel et al. [14], Pujana-arrese at al. [17] and Wickramatunge and Leephakpreeda [21] or in Hildebrandt et al. [10] where the polynomial is mixed with exponentials. However these models are not trivially invertible with good accuracy (see [3]) so they cannot be used for control purpose, where inverse models are usually used to deal with

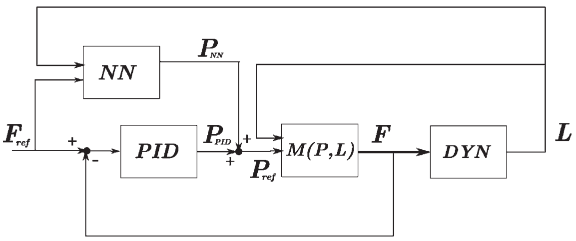

Fig. 7. The control system block diagram. The reference pressure to the servovalve is computed as $P_{ref} = P_{NN} + P_{PID}$, where $P_{NN} = NN(F_{ref}, L_m)$ is the neural network contribution and $P_{PID} = PID(F - F_{ref})$ is the PID contribution.

non-linearity. To obtain an inverse model of actuator we propose a black box approach using neural networks. To do this we trained the network reversing the muscle inputs and outputs.

Figures 8 and 9 shows the performance of the proposed controller whose structure is reported in Fig. 7. Muscles couple with the mechanical system through force transmission F and feedback L. This feedback to the muscle describes the fact that force depends on the length, which in turn depends on system dynamics so the feedback line can describe this relation.

In Fig. 10 we show the controller performance during a treatment with a real patient. Here the force trajectory is smoother because of slower dynamics of the leg system has the above mentioned filtering effect. With patient wearing the orthosis we reach a maximum dynamic error of 1.6 N while in quasi steady state condition (is quite difficult to obtain really static condition because the patient is always moving a bit) the error is usually less than about 1.0 N.

5. Results

As final validation of our prototype we organized some walking sessions with a 17 years old male patient. Figure 11 shows how he wears the orthosis. Firstly we test the system in a passive configuration, setting very low and constant force reference, just to keep pneumatic muscles stretched. We noticed that actuators were able to follow human movements without obstruct them. Then we try gradually to test the force generation algorithm. The final result was evident. From the beginning the user was able to move in a quite comfortable way and being familiar with prototype he could reach durable and autonomous walking.

In previous passive orthosis treatment - lasting two years - our test patient always needed some help from physiotherapist. Aid can be by hands or by sticks. In the first case patient need to be pulled by hand reaching a comfortable velocity that let him to make some steps in almost autonomy. However, after few steps he quickly loses kinetic energy and halts after six or seven steps. The second way to help patient is manually forcing the gait, this can be done acting on the reciprocation bends through small sticks. In this case the patient can walk for longer period, but in an almost passive way.

At the very beginning patient behaviour with ARGO was very similar to the help by hand. He needed to reach a threshold velocity to trigger a continuous walking mechanism. On the other hand, ARGO actuation tries to replicate the help by sticks. The main difference is that artificial muscles are activated by the patient himself rather than by an external helper. In particular,

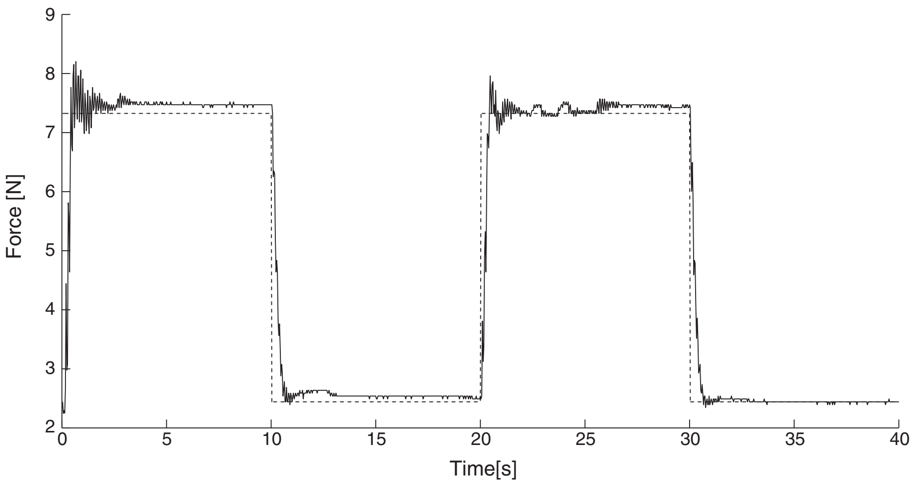

Fig. 8. The proposed controller on square wave trajectory. Dashed line is the reference.

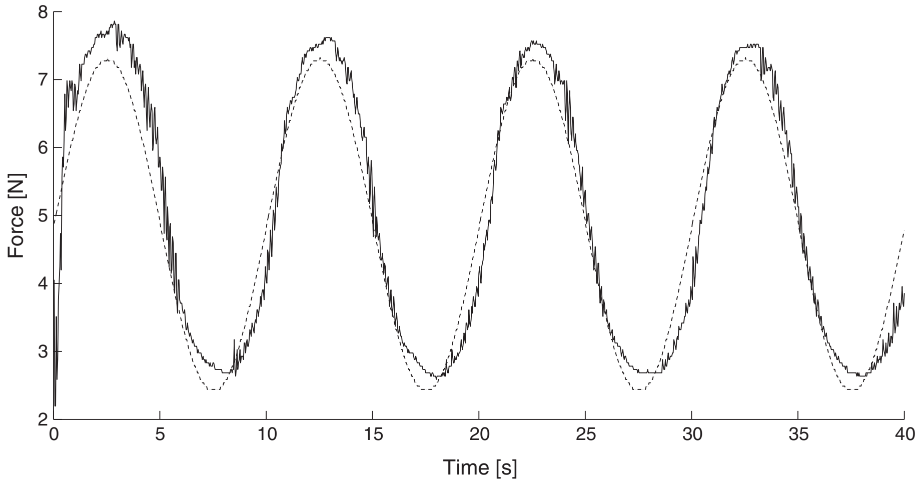

Fig. 9. The proposed controller on 0.1 Hz sine trajectory. Dashed line is the reference.

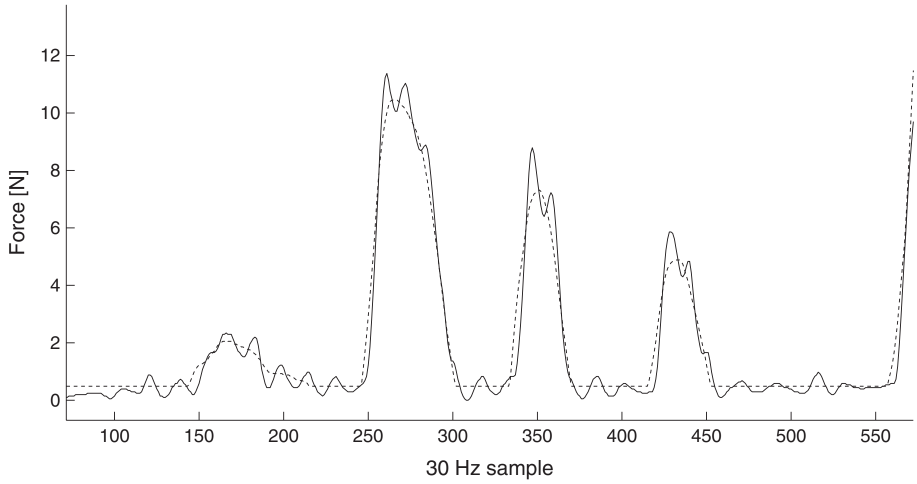

Fig. 10. The proposed controller on a real treatment trajectory. Dashed line is the reference.

PAMs are triggered mainly by ground reaction force distribution over feet. Training with orthosis the patient learns to deploy this mechanism and begins to trigger the walking by himself. After about five hour of total training - divided into eight sessions - he was able to perform a long sequence of steps without external help.

We can say that the prototype role is to supply power in a way that can be controlled by the patient, who needs to learn how to deploy this effort. In this exchange the orthosis contributes to the bio-feedback mechanism of the patient like a body extension that provides patient missing power or that amplifies the sense of movement. However, through EMG analysis (Fig. 12), we do not notice significant differences in muscle recruiting. On one side this means that the active orthosis does not make the user passive. This thesis is supported also by energetic consumption analysis. Using Cosmed K4 system, we found that the fatigue is not reduced, as showed in Table 1. The energy cost of walking is determined from the ratio of the net V_{O_2} to the speed of progression. The net O_2 consumption is obtained by subtracting the oxygen needed for basal metabolism in a resting sit-down posture. The net consumption provides the additional oxygen required for anti-gravity support and propulsion (see Schwartz et al. [18] for details on energy cost).

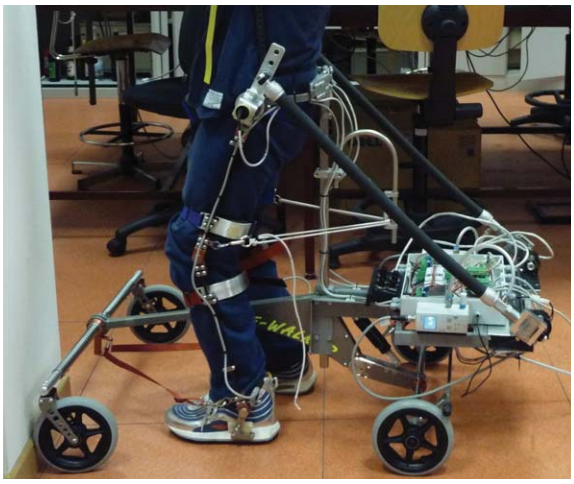

Fig. 11. ARGO worn by our test patient.

In conclusion we can think that the overall effect of actuation is to couple with patient muscle, changing the orthosis-legs system dynamics in a way that seems easier controllable. Figure 14 shows a snapshot progression of an experimental session with the patient and Fig. 13 shows an example of data collected during such experiments. We plot the torque profile against hip position for both the left and right leg. It is easy to see that to each step corresponds a different force contribute, depending on patient behaviour. The bottom part graph shows that the user is free to perform a double left step and the system easily understands his intention. Data from the FSM are also reported as a validation of the gait phase recognition.

6. Conclusions and future works

This article presents a detailed description of an active orthosis prototype targeting Cerebral Palsy children that we named ARGO. ARGO's structure is based

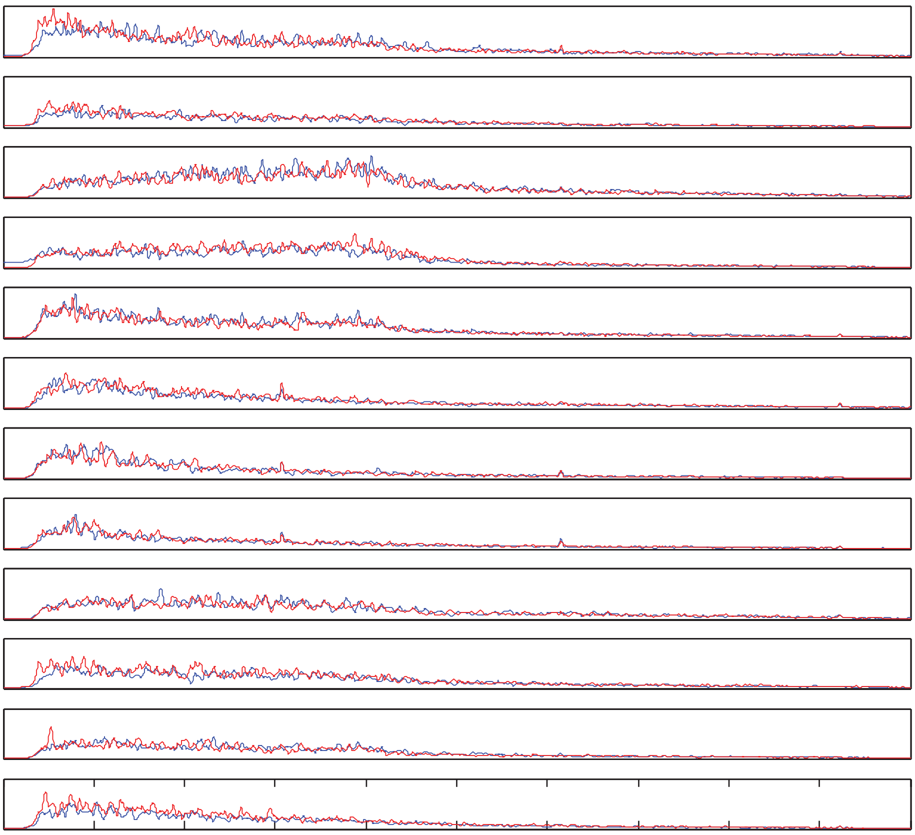

Fig. 12. Subplots are relative to the power spectral density of EMG signals from biceps femoris R and L, rectus femoris R and L, tensor fascia lata R and L, gluteus maximus R and L, vastus lateralis R and L, semitenosus R and L. Ordered from the top plot to the bottom one. Frequency is expressed in Hz.

Table 1
Energetic Cost with active (A) and passive (P) orthosis. V is the average velocity, V_{O_2} is the steady state oxygen uptake and EC is the energy cost considering a basal of 2.69 $J/Kg/m$

	V m/s	net V_{O_2}/Kg ml/min	EC $J/Kg/m$
P	0.18	22.6	**38.5**
A	0.15	18.6	**36.1**

on a commercial orthosis to take advantage of the mechanical optimizations and extensive user's testing maturated over time.

The actuation system is patient controlled, using a force-based control algorithm. It uses PAM in non-antagonist muscle configuration, deploying mechanical reciprocation. This keeps the system lighter and reduces air consumption. To recognize the gait phase we built a finite state machine which analyses and relates data from the hip position, ground reaction forces and patient-actuator interaction forces. Basing on the gait phase state a force planning algorithm generates hip torques.

On a single case study we have observed that the prototype has some benefits. Our patient was not able to walk by himself after two years of training with the passive orthosis while after some hours of training with the prototype he achieved autonomy. Up to now

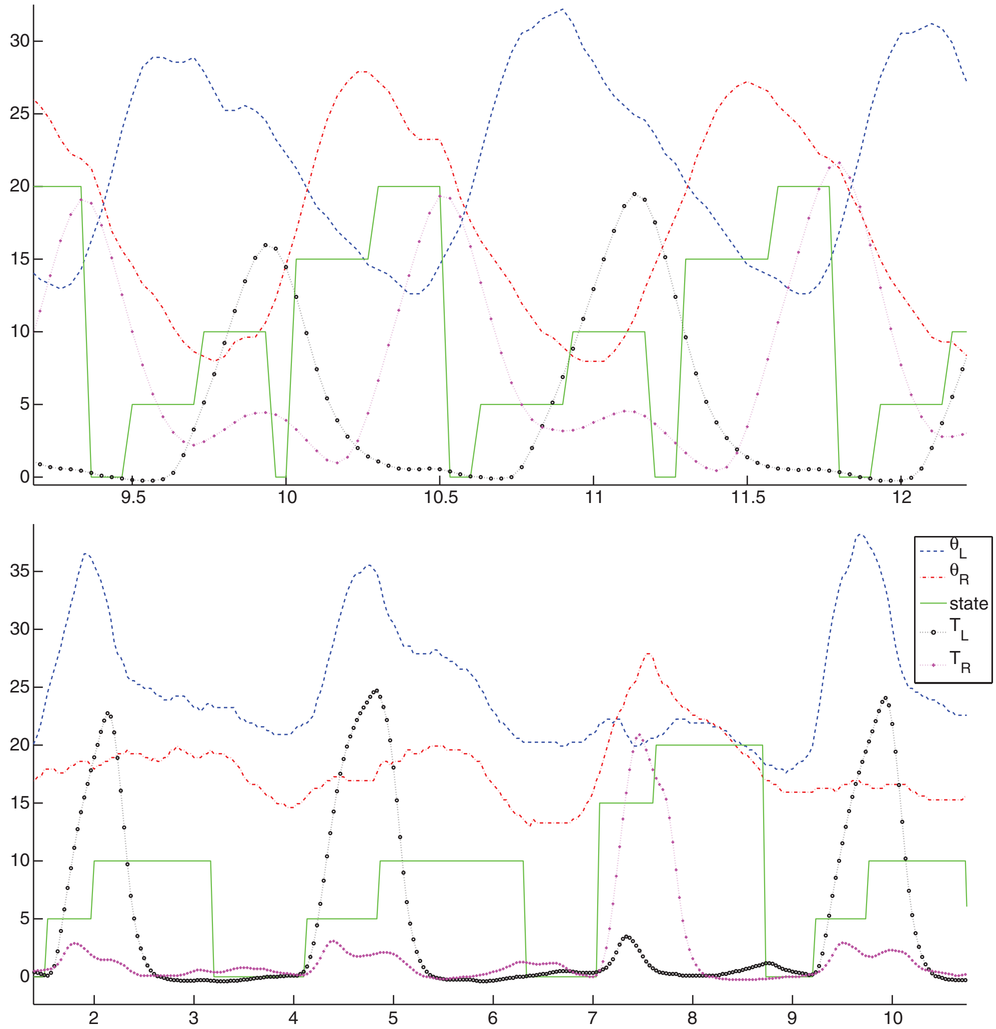

Fig. 13. ARGO contribution to user's gait. The solid line represents the gait phase according to the states in Fig. 2 where SS, LU, LD, RU and RD correspond respectively to 0, 5, 10, 15 and 20 value. The markers represent the hip joint angle [°], while dashed and the dot-dashed lines represents the torque applied to the ankle, in scaled units [$\frac{1024}{500}$ Nm].

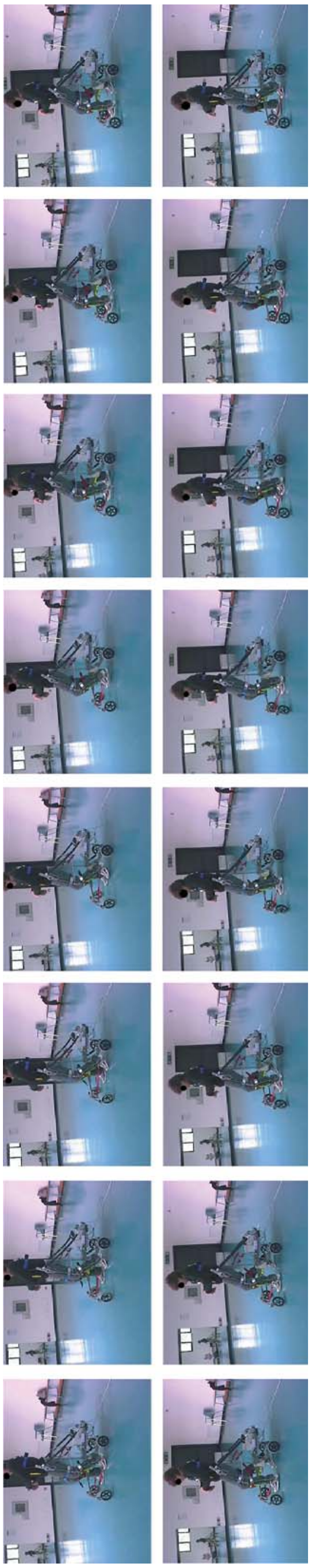

Fig. 14. Sequential snapshots collected during an experiment session.

there is not clinical evidence of rehabilitation robotics benefits for CP. This issue is actually not treated in literature because of two basic reasons: young patients brain is rapidly evolving and neural lesion are usually more complex than other pathologies such as stroke and spinal cord injury. Due to these difficulties CP patients are usually excluded from clinical study.

On the basis of the experimental results we are actually looking for replicate the prototype (with some major modification) to get long period data from different patients. Together with a clinical staff this can lead us to begin to better understand something about robotic rehabilitation and physical therapy for CP.

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